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Washing and Abrasion Resistance of Textile Electrodes for ECG Measurements

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Abstract: Electrocardiogram (ECG) signals are often measured for medical purposes and in sports. While common Ag/AgCl glued gel electrodes enable good electrode skin contact, even during movements, they are not comfortable and can irritate the skin during long-term measurements. A possible alternative is textile electrodes, which have been investigated extensively during the last years. These electrodes, however, are usually not able to provide reliable, constant skin contact, resulting in reduced signal quality. Another important problem is the modification of the electrode surface due to washing or abrasion, which may impede the long-term use of such textile electrodes. Here, we report a study of washing and abrasion resistance of different ECG electrodes based on an isolating woven fabric with conductive embroidery and two conductive coatings, showing unexpectedly high abrasion resistance of the silver-coated yarn and optimum ECG signal quality for an additional coating with a conductive silicone rubber. Sheet resistances of the as-prepared electrodes were in the range of 20–30 Ω, which was increased to the range of 25–40 Ω after five washing cycles and up to approximately 50 Ω after Martindale abrasion tests. ECG measurements during different movements revealed reduced motion artifacts for the electrodes with conductive silicone rubber as compared to glued electrodes, suggesting that electronic filtering of such noise may even be easier for textile electrodes than for commercial electrodes.

Keywords: electrocardiogram (ECG); textile electrodes; conductive coating; conductive yarn; sensor; Arduino

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1. Introduction

Since the first development of smart textiles, textile electrodes for electrocardiogram (ECG) measurements have emerged as the most frequently investigated objects [1–3]. ECG measurements can help to identify heart-related health issues, such as arrhythmias, heart attacks, heart failure, heart defects, etc., which are the most frequent causes of death nowadays in many parts of the world [4,5].

Stationary ECGs are usually measured as 12-lead ECGs, consisting of the six precordial leads according to Wilson (using electrodes on the chest, from approximately the middle of the front to a position near the left axilla), three limb leads according to Einthoven (with electrodes located at the left arm, right arm, and left leg), and three augmented limb leads according to Goldberger (using the same electrodes as Einthoven, combined with a common virtual electrode which is calculated from averaging the previously mentioned three limb electrodes), which together enable the detection of the electrical processes in the heart from diverse directions [6]. Such full 12-lead ECG systems, however, are only infrequently integrated into textiles [7]. Instead, most textile-based ECG measurements are performed using three leads, often integrated into clothing similar to the positions according to Einthoven [8–10]. This positioning is due to the fact that suitable pressure on the chest electrodes is hard to achieve by a textile fabric, especially in the case of female probands, and that the augmented limb leads according to Goldberger need more computation. The positions at the arms and the leg can be shifted arbitrarily towards the heart, as long as the measurement orientation relative to the heart is maintained, making it relatively easy to find suitable positions where sufficient pressure can be exerted on textile-included electrodes.

Among the main problems of textile-based electrodes, the low skin contact has to be mentioned. In contrast to glued gel electrodes, which reduce the electrical contact resistance between the metal part of the electrode and the skin by a conductive gel, textile electrodes only partly touch the skin without any bridging medium and thus usually show much higher contact resistance. This problem is usually aimed to be solved by increased pressure on the electrodes towards the skin and by using drapable electrodes, which can follow the body contours [11–13]. Conductive coatings with low water vapor permeability can additionally improve the skin contact by providing a fine layer of sweat between electrode and skin [14–16]. Besides improving the textile part of the measurement equipment, sophisticated electronic filters can help reducing the 50/60 Hz interference as well as noise from body-based signals, such as moving or breathing [17–19].

However, garment-integrated textile electrodes need to be washed regularly and will experience abrasion during their lifecycle. These factors influencing the potential long-term use of textile ECG electrodes have been investigated by some research groups. Ankhili et al. found a substrate-dependent increase of the sheet resistance of textile electrodes prepared from poly(3,4-ethylenedioxythiophene):poly(styrene-sulfonate)(PEDOT:PSS)-coated cotton, polyamide, or polyester fabrics [20]. Arquilla et al. found a clear increase in the resistance of electrodes with sewn silver-coated threads over eight washing cycles [21]. Soroudi et al. used silver-coated Shieldex yarn, partially with additional silver coating, and found reduced motion artifacts of the latter [2]. Wang et al. coated thiol-group grafted polyester fabrics with a condensed silver layer by electroless plating and found a good washing resistance for the conductivity of these special conductive fabrics [22].

In addition to metallization and coating with conductive polymers, textile fabrics can also be made conductive by carbon-based coatings, e.g., with carbon black or graphite. One of the commercial carbon black coatings, Powersil, is often used as a textile coating, but the effects of washing have only scarcely been investigated [23–25], and it has not yet been tested with respect to using ECG electrodes after washing or abrasion.

Here, we report a comparison of textile ECG electrodes based on a woven cotton fabric with embroidered metalized yarn, Powersil, and PEDOT:PSS coating after washing and abrasion. Measurements of the sheet resistance of the three types of electrodes are followed by ECG measurements of the as-prepared electrodes in comparison with commercial glued electrodes. For the Powersil-coated electrodes, which are found to be advantageous for ECG measurements, further tests are performed during different states of movement, comparing the ECG measurements obtained with these textile electrodes with the results of ECG measurements with glued electrodes.

2. Materials and Methods

The base material for all textile electrodes was a jeans (100% cotton) fabric, chosen due to its robustness and since it could withstand the oven temperature of 200 °C needing for one coating step.

For conductive yarn, the silver-coated yarn Shieldex 235/34 dtex 2-ply HC+B (Statex, Bremen, Germany) was used. Conductive coatings were Powersil 466 A/B (Wacker Chemie AG, München, Germany), a liquid silicone rubber based on polydimethylsiloxane, and PEDOT:PSS (Clevios S V 4, Sigma-Aldrich Chemie GmbH, Munich, Germany), a conductive doped polymer.

Embroidering of the conductive yarns on the jeans fabric was performed using the sewing machines W6 N1800 (W6 WERTARBEIT Projektierungs- und Handelsgesellschaft mbH, Wennigsen, Germany) and Juki HZL-DX3 (Juki, Röhrsdorf, Germany). The stitches used were backstitch + zigzag stitch with a step width of 2.5 mm, identical to those in a previous study on Shieldex-embroidered bioimpedance electrodes [23]. The previous experiments found that this kind of stitch combination ideally suited for textile electrodes [23], which is why it was also used here. Powersil coating was applied by a doctor's blade, followed by polymerization for 2 h at 200 °C. PEDOT:PSS coating was performed by dipcoating, followed by drying for 4 h at 60 °C. The following three types of electrodes were produced:

- Conductive sewing only (Figure 1a): cutting an electrode substrate from cotton, sewing the borders with non-conductive yarn to avoid unraveling, sewing three lines of backstitch + zigzag stitch, as described above.
- Conductive sewing followed by PEDOT:PSS coating (Figure 1b): starting with the aforedescribed electrode with conductive sewing → dipcoating with as-purchased PEDOT:PSS → 4 h at 60 °C.
- Conductive sewing followed by Powersil coating (Figure 1c): starting with the aforedescribed electrode with conductive sewing → mixing components A and B of Powersil in a ratio of 1:1 → coating the electrodes using a doctor's blade → polymerization for 2 h at 200 °C.

(**a**) (**b**) (**c**)

Figure 1. Example electrodes, made conductive (**a**) by conductive sewing only; (**b**) by conductive sewing followed by PEDOT:PSS coating; (**c**) by conductive sewing followed by Powersil coating.

To investigate washing fastness, 5 washing cycles were performed in a household washing machine at 40 °C with Frosch (Werner & Mertz GmbH, Mainz, Germany) heavy duty detergent. Abrasion resistance was investigated by a custom-made Martindale abrasion tester according to ISO 12947-1:1998 [26] up to 7000 abrasion cycles.

Sheet resistance measurements were performed with a 4-point measurement system MR1 (Schuetz Messtechnik, Teltow, Germany), measuring along the sewing direction on the conductive yarns for those electrodes where the conductive yarn is openly visible, and measuring along the whole surface including near the edges in case of the Powersil-coated electrodes. ECG measurements were obtained using an AD8232 (Analog Devices, Wilmington, MA, USA) ECG sensor module (Sparkfun, Niwot, CO, USA) attached to an Arduino Uno, displaying the measurement in the monitor of the Arduino IDE in a sketch suggested by Sparkfun [27]. The three electrodes were placed on a proband's chest, as described by Einthoven, and slightly pressed onto the skin by wrapping them with an elastic textile fabric.

3. Results and Discussion

Sheet resistances were measured along the embroidered parts of the electrodes depicted in Figure 1. It should be mentioned that the test electrodes of the 4-point measurement system did not always remain in full contact with sewn stitches, which either resulted in a measurement error that was not taken into account, or in a modified measurement result in case of electrodes with conductive sewing and the less-conductive PEDOT:PSS coating, resulting in a lower result and, finally, in larger standard deviations. The Powersil coating is generally less conductive than the silver-coated yarn, so these electrodes can be expected to show higher sheet resistance values. Due to the aforementioned potential measurement errors, more than 100 single measurements per electrode were taken on 4 nominally identical specimens per sample.

The results of the measurements before washing and abrasion are depicted in Figure 2. As expected, large error bars arise from the uneven structure of the textile surfaces, especially in the case of the purely embroidered electrodes, which impede an always identical contact area between textile surface and measurement electrodes. In addition, the Powersil coating is of uneven thickness due to the coating process on the previously embroidered electrodes, leading to position-dependent conductivity. Nevertheless, all values can be estimated to be suitable for use as textile electrodes since resistances in the range of tens of Ohms are typical for textile ECG electrodes [12,28,29], and even electrodes with resistances in the range of kilo-Ohms have been shown to be suitable for ECG measurements [15]. The differences between the three types of electrodes are small and not significant.

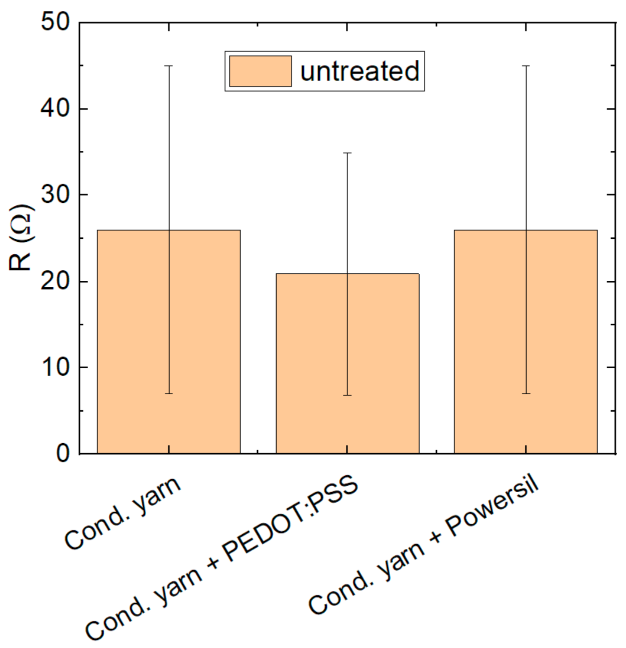


Figure 2. Sheet resistances of untreated electrodes.

Next, the electrodes were washed, and sheet resistance values were measured after one washing cycle and after five washing cycles. The results are depicted in Figure 3.

For all three kinds of electrodes, a tendency towards higher sheet resistance values after washing is apparent. While this increase is obviously not significant, as shown by the large error bars, it corresponds to the expected trend. However, it must be mentioned that the average sheet resistance values are increased by less than 50% after five washing cycles, while Powersil coatings on different textile fabrics usually show an increase in resistance by a factor of five to ten after five washing cycles [14], suggesting that washing will not significantly alter these electrodes' ability to detect ECG signals.

Besides washing, the abrasion resistance was investigated. The results after Martindale abrasion tests (pure sewing after 7000 Martindale cycles, PEDOT:PSS-coated electrodes after 400, and Powersil-coated electrodes after 3000; numbers correspond to maximum test cycles according to optical examination, as explained below) are depicted in Figure 4, in comparison with the original electrodes.

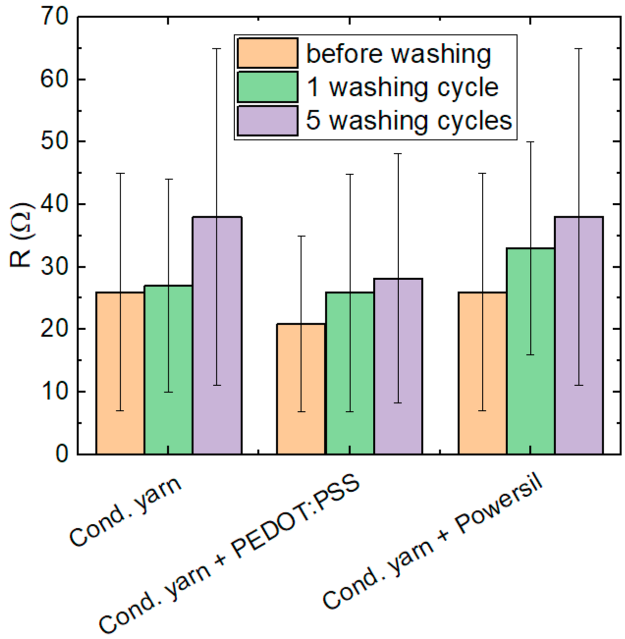


Figure 3. Sheet resistances of electrodes after different numbers of washing cycles.

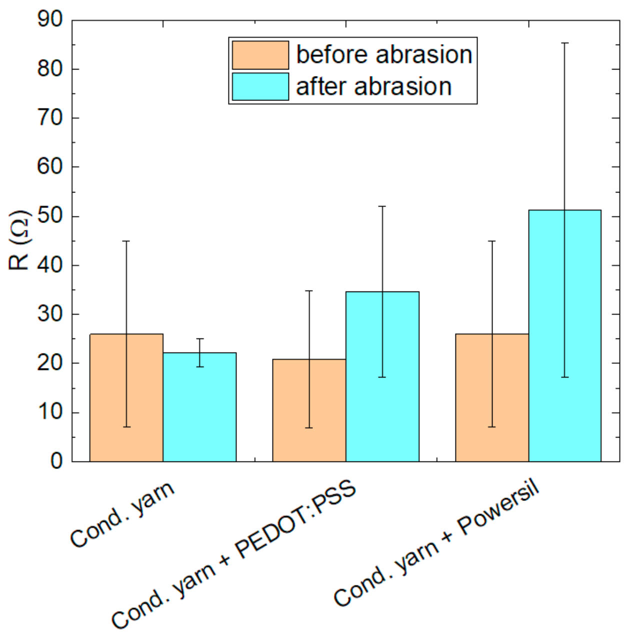
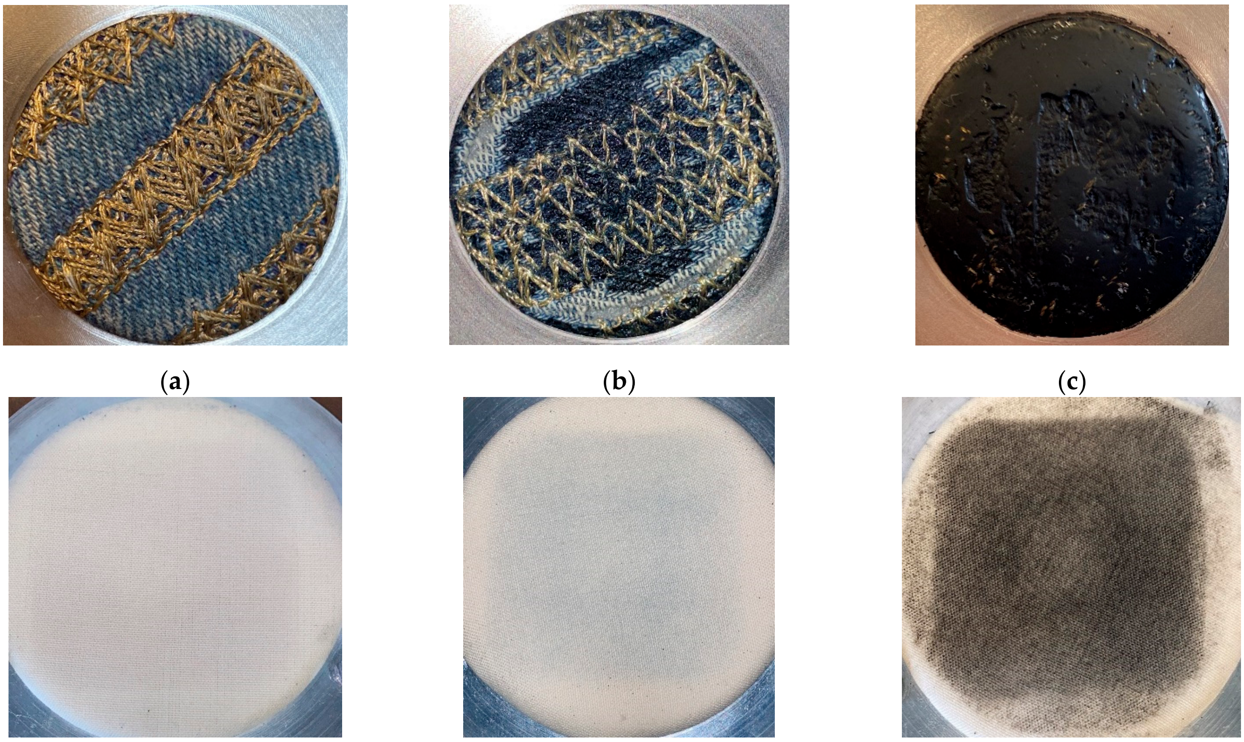


Figure 4. Sheet resistances of electrodes before and after Martindale abrasion tests.

Unexpectedly, abrasion tests on the electrodes with pure sewn yarns did not increase their resistance. This is different for both coated samples, with the electrodes with Powersil coating showing the largest resistance after abrasion tests. Nevertheless, even these sheet resistance values around 50 Ω are still uncritical for ECG measurements.

In addition to the electrical tests, the electrodes were also investigated optically during the abrasion tests. Example results are depicted in Figure 5. The electrodes with only embroidered yarn remained stable after 7000 Martindale abrasion cycles (Figure 5a); the test was stopped then since garment-integrated textile electrodes will not experience such harsh treatment during application. No coloration of the fabric woven against it was abraded was visible after the test (Figure 5d). Unexpectedly, the electrode with additional PEDOT:PSS coating was destroyed already after 400 Martindale cycles (Figure 5b), where, in particular, the jeans fabric was clearly damaged. Since these electrodes were only treated at 60 °C in the oven, no thermal damage can have occurred; instead, the PEDOT:PSS coating must either have damaged the cotton or increased the friction with respect to the standard abrasive woven fabric. In both cases, these electrodes may be damaged also during application in an ECG measuring garment. The abrasive woven fabric shows slight coloration (Figure 5d), whether from the dyed cotton or from the PEDOT:PSS is not clearly visible. Finally, the Powersil-coated electrode loses its gloss (Figure 5c), and strong coloration of the counteracting abrasive woven fabric is visible (Figure 5f).



(**a**) (**b**) (**c**) (**d**) (**e**) (**f**)

Figure 5. Martindale abrasion tests: (**a**) Pure conductive sewing after 7000 cycles; (**b**) sewing + PEDOT:PSS coating after 400 cycles; (**c**) sewing + Powersil after 3000 cycles; standard abrasive woven fabric after tests with (**d**) pure conductive sewing; (**e**) sewing + PEDOT:PSS coating; (**f**) sewing + Powersil after the aforementioned numbers of abrasion cycles.

While the previous tests suggest excluding the PEDOT:PSS coated electrodes from application in smart textiles for ECG measurements, it is also necessary to measure ECG signals with all electrodes under examination. Representative results are shown in Figure 6. Since the scales of the Arduino serial plotter are very small, the y-ranges for all plots are given in the figure caption. All x-axis spans are 5 s.

Using the electrodes with pure conductive sewing (Figure 6a), no ECG signals could be detected in the recent setup, i.e. without exerting much pressure onto the skin to improve the electrode–skin contact. The additional PEDOT:PSS coating (Figure 6b) sometimes allowed a signal to be recognized, but was always associated with a saturated signal, so this material combination was also not suitable. Only the Powersil coating (Figure 6c) led to an increase in skin contact so that the QRS complexes, i.e., the most prominent peaks of the full ECG signal, could be detected. The signal is still not absolutely clear due to using unshielded cables as well as breathing and moving during the measurements (which is filtered in professional systems), but could be used for a simple long-term pulse detection. The signals are slightly noisier than those taken with commercial glue electrodes (Figure 6d), but this can be amended by increasing the pressure on the skin.

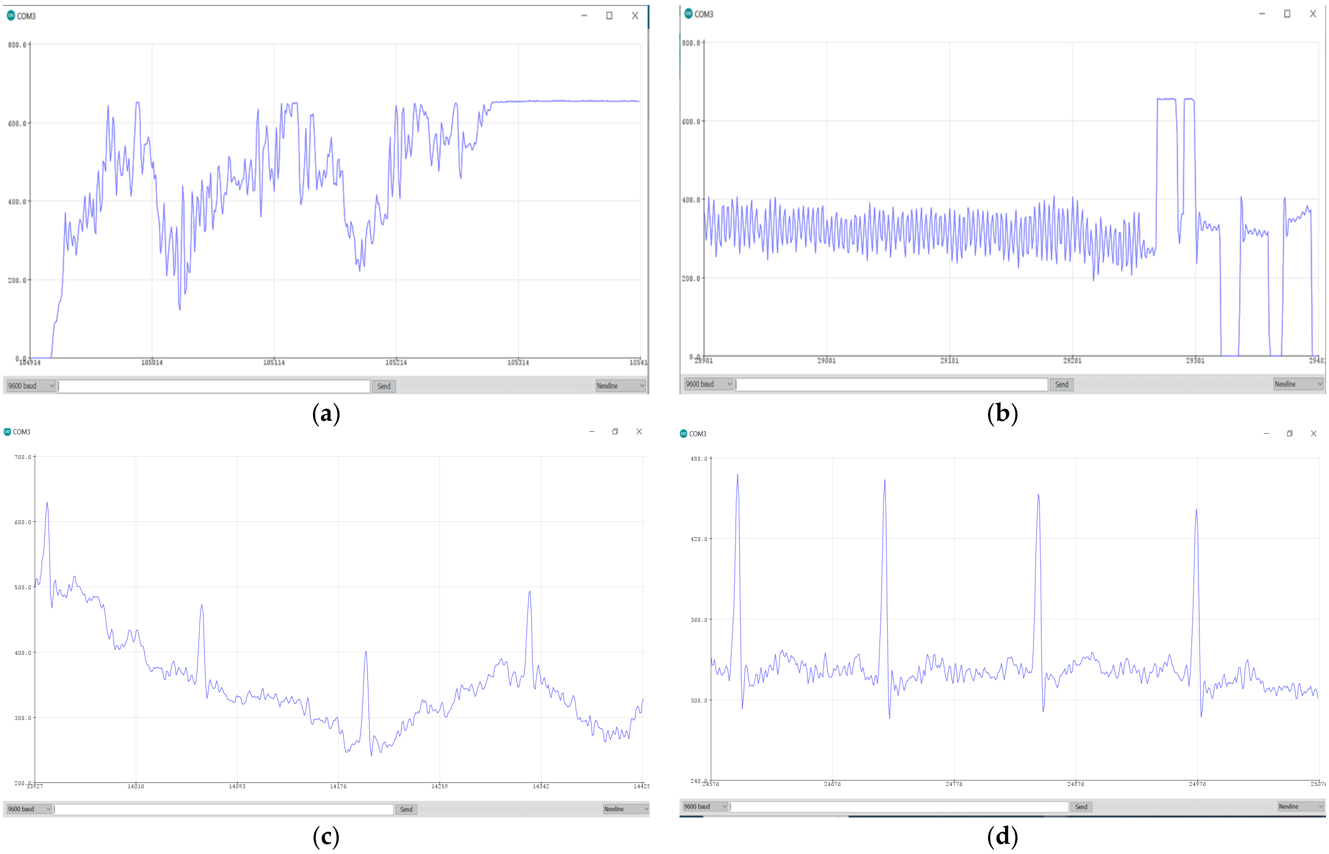


Figure 6. ECG measurements with textile electrodes: (**a**) pure conductive sewing (y-scale 0–800); (**b**) sewing + PEDOT:PSS coating (y-scale 0–800); (**c**) sewing + Powersil (y-scale 200–700); (**d**) measurement with commercial glue electrodes (y-scale 240–480).

The difference between Powersil-coated and other electrodes regarding the ECG measurement is, on the one hand related, to the surface of the coating. The soft, compressible, rubber-like coating builds a good skin contact along the whole electrode area, whereas the electrodes with conductive embroidery only make electrical contact with the skin along the single conductive yarns, i.e., the contact area is much lower, leading to increased noise upon the smallest movements of the electrodes relative to the skin and decreased signal due to higher contact resistance. This problem is not solved by the PEDOT:PSS coating, as this low-viscosity coating fluid sinks into the fabric and provides a higher conductivity parallel to the surface, but not perpendicular to it. On the other hand, the Powersil coating has a very low water vapor permeability, resulting in a fine layer of sweat building on the skin and further improving the contact resistance. Since the pressure of the electrodes on the skin is very low here, unlike in previous studies [7,8], the latter effect is especially important since the sweat layer closes the thin gaps that otherwise occur between the "valleys" in the skin and the electrode, keeping in mind that both parts of the contact are not perfectly flat.

As these measurements show, Powersil-coated cotton-based electrodes with additional conductive threads to reduce the in-plane resistance are not only suitable for ECG measuring, but can also withstand washing and even abrasion over a long period and can thus be used for integration into smart garments, measuring pulse or ECG in the long-term monitoring of elderly people or in clothes for sports.

To investigate these electrodes further, Figure 7 depicts measurements with the five-times washed Powersil electrodes in comparison with glued electrodes for the situations of "no movement" (i.e. low breathing) and "heavy breathing". The same electrodes are compared in Figure 8 for the situations of "running" and "arms moving up and down at the sides of the body", respectively.

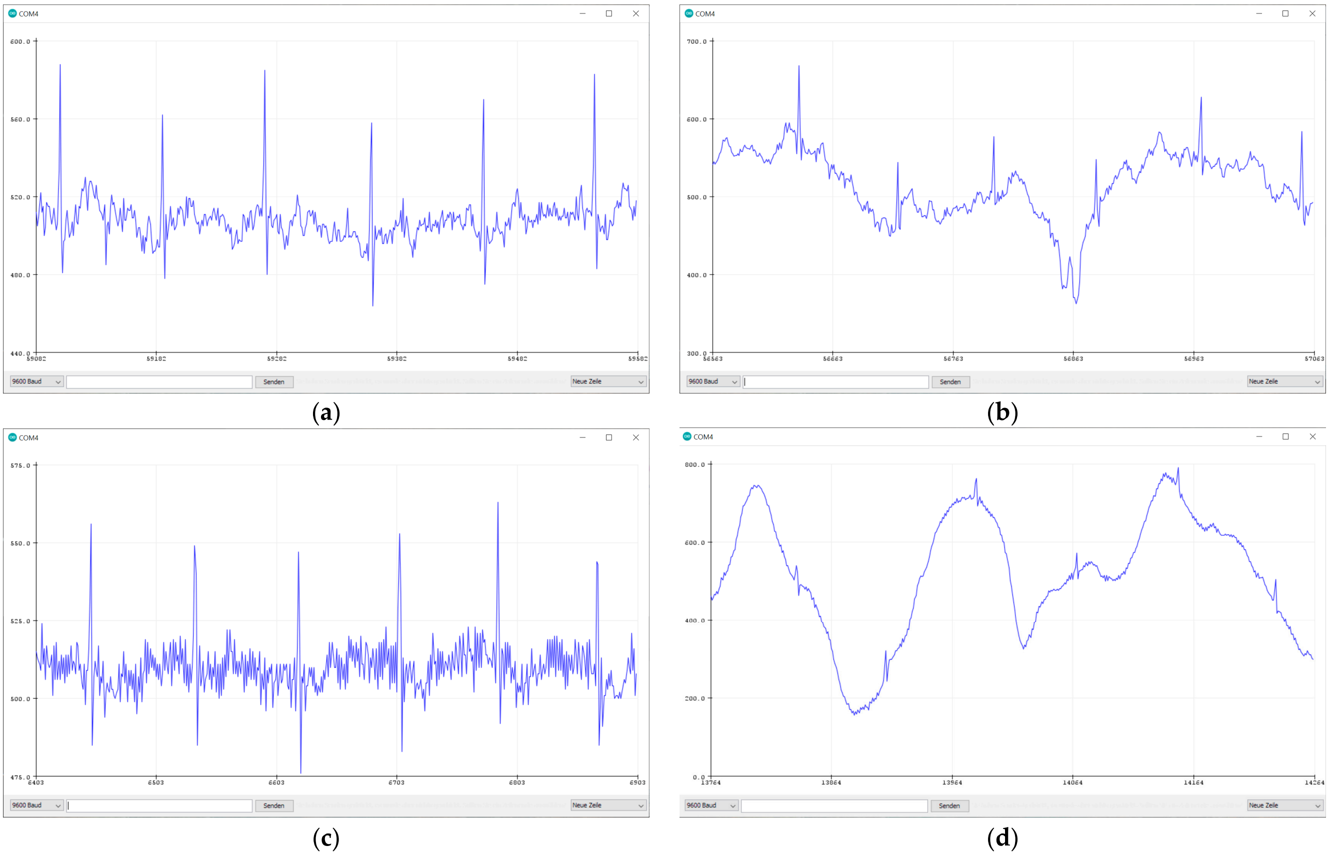
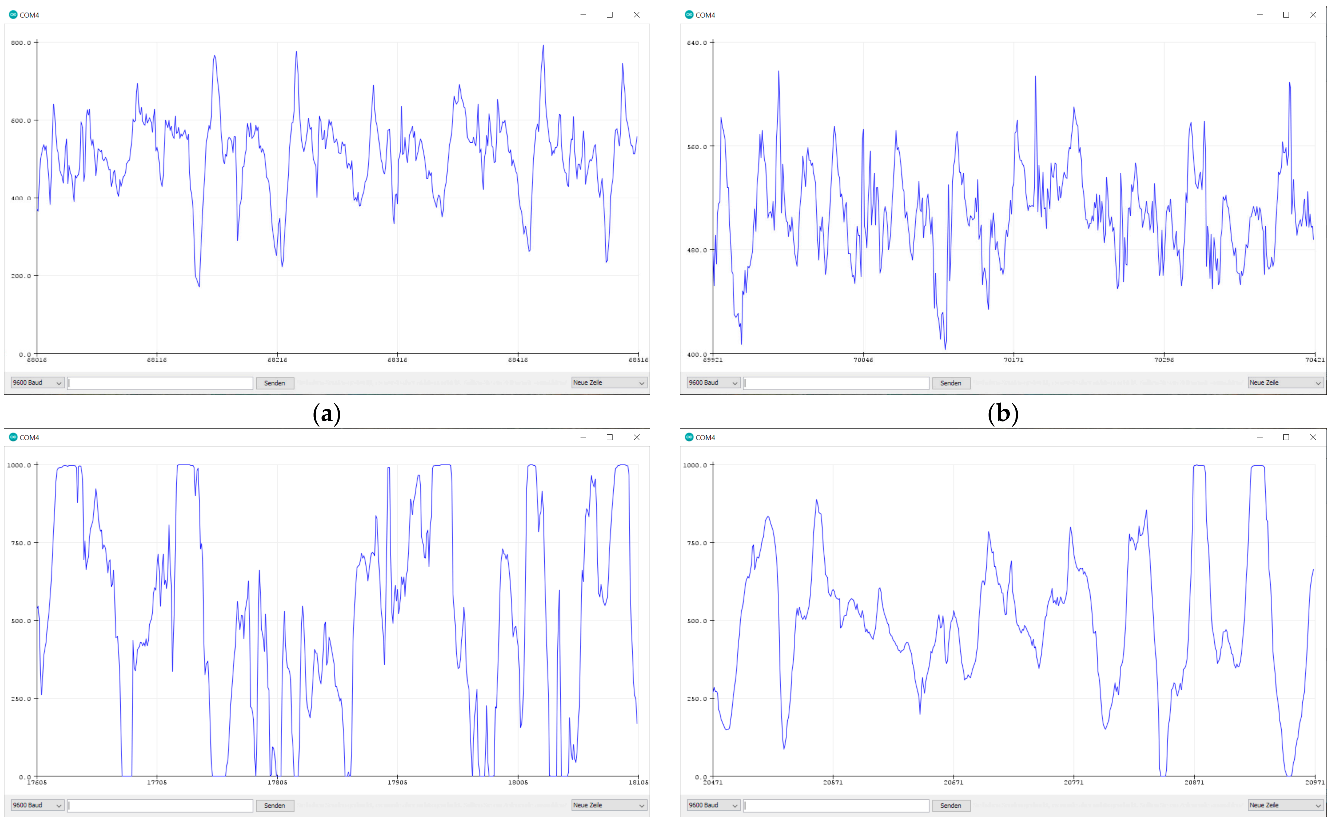


Figure 7. ECG measurements with washed Powersil-coated electrodes: (**a**) no movement (y-scale 440–600); (**b**) heavy breathing (y-scale 300–700); compared with glued electrodes: (**c**) no movement (y-scale 475–575); (**d**) heavy breathing (y-scale 0–800).

During rest, both the Powersil-coated (Figure 7a) and the glued electrodes (Figure 7c) show the QRS complexes in the form of spikes. The T-waves that follow the QRS complexes are hardly visible, while the P waves which precede the QRS complexes are invisible. It should be mentioned that the recent measurements are based on low-cost equipment that will be developed further in the near future. The aim of the recent project is not improving the electronics and adding filters to reduce noise from the environment or from movement artifacts, but investigating which textile electrodes have the largest potential to be used in combination with low-cost, lightweight electronic equipment which can be embedded in textile garments and worn all day long without restricting the patient's freedom of movement. The noise visible here is thus ignored in the recent study.

Comparing the measurement with the washed Powersil electrodes (Figure 7a) with the results of the as-prepared Powersil electrodes (Figure 6c), no differences are visible, showing that the five washing cycles do not reduce the quality of the Powersil-coated electrodes regarding their usability as ECG electrodes.

Figure 7b and 7d show measurements of ECG signals during heavy (deep) breathing, obtained with Powersil-coated (Figure 7b) and glued electrodes (Figure 7d), respectively. Interestingly, the large "waves" on the signal visible for the glued electrodes (Figure 7d) due to the body movements during breathing are much higher than those visible for the measurement with the Powersil-coated electrodes (Figure 7b). This unexpected finding may be based on the larger electrode area of the textile electrodes compared to the glued electrodes, which averages the measured signal over a larger area and thus potentially reduces the impact of motion artifacts.



(**a**) (**b**)

(**c**) (**d**)

Figure 8. ECG measurements with washed Powersil-coated electrodes: (**a**) running (y-scale 0–800); (**b**) arms moving up and down at the side of the body (y-scale 400–640); compared with glued electrodes: (**c**) running (y-scale 0–1000); (**d**) arms moving up and down at the side of the body (y-scale 0–1000).

The measurements taken during stronger movements, as depicted in Figure 8, show similar effects. During (slow) running, the arms of the proband performed the typical movement back and forth along the sides of the body. The results are depicted in Figure 8a for Powersil-coated electrodes and in Figure 8c for glued electrodes. Similar to the previous measurements during deep breathing, again the motion artifacts are stronger visible in the measurement with the glued electrode (Figure 8c), where both upper and lower saturation are reached. For the arm movement up and down at the sides of the body, as depicted in Figure 8b for the Powersil-coated electrodes and in Figure 8d for the glued electrodes, the same effect can be recognized, but only the signal from the glued electrodes reaches both saturation values. As discussed before, it can be hypothesized that the larger areas of the textile electrodes help to reduce motion artifacts to a certain extent. This finding suggests that the planned electronic filters will work better for textile electrodes than for glued electrodes.

To compare the results of our study with other recent literature reports, Table 1 presents an overview of the key parameters of these studies. Usually, these studies use commercial ECG devices or self-built devices with electronic filters for 50 Hz noise and motion artifacts and use static measurements, although a few [16] also perform tests during running, when the body produces sweat, which strongly increases the skin contact. The electrodes produced in our study show an average sheet resistance and were the only one to reduce motion artifacts as compared to common glued electrodes. Consequently, removing the usual filters can be considered an advantage of the new ECG electrodes.

Table 1. Key parameters of recently reported textile ECG electrodes.

Textile Structure	Conductive Materials	Sheet Resistance/Ω	Main Features	Test Conditions	Reference
Woven	Copper/copper nickel coating	0.04	Washing impossible	Notch filter + bandpass filter, probands rested before tests	[28]
Knitted	Silver-plated	0.3–1.5	Washing possible		[28]
Knitted	Screen-printed silver ink	1.6–1.8	Pressure dependent signal	Commercial ECG device, probands rested before tests	[12]
Woven	Screen-printed graphene	42	Bending-resistant	Pre-amplifier, filter, post-amplifier	[29]
Warp-knitted	Silver-plated	0.1–1.7	Lower signal height than with gel electrodes	Clinical 1-lead ECG system	[30]
Woven	PEDOT:PSS screen printing	330	Signal similar to gel electrodes	Static measurement	[31]
Woven	Graphene oxide dyeing by followed by PEDOT:PSS coating	50×10^3	Similar to medical electrode	Clinical ECG monitor, tested during running	[16]
Knitted	Reduced graphene oxide and PEDOT:PSS coating	140×10^3	Measured at the wrist joints	Clinical ECG monitor	[32]
Woven	Silver-coated yarn with Powersil-coating	20–30	Washing possible, lower motion artifacts than gel electrodes	Low-cost system without filters	This work

4. Conclusions

Textile electrodes were produced by sewing with conductive silver-coated yarns, partly with additional dip-coating with PEDOT:PSS or doctor blade coating with Powersil. The sheet resistance values were similar for all three types of electrodes and not notably increased during washing or Martindale abrasion tests. Unexpectedly, the PEDOT:PSS-coated jeans fabric was destroyed after only 400 Martindale cycles, while both other fabrics withstood several thousand cycles. Sheet resistance values were in the range of 20–30 Ω for the as-prepared electrodes, slightly increased to 25–40 Ω after five washing cycles, and reached about 50 Ω after Martindale abrasion tests.

In the ECG measurements, only the Powersil-coated electrodes resulted in ECG signals; neither of the other types of electrodes had sufficient skin contact under slight pressure to enable ECG measurement. The Powersil-coated electrodes were able to measure ECG signals after five washes with the same quality as in the as-prepared state. Unexpectedly, all measurements taken during different movements revealed the lower impact of motion artifacts with the Powersil-coated electrodes as compared to commercial glued electrodes.

Our results suggest further investigation of Powersil-coated ECG electrodes regarding the optimum pressure on the skin and the best electrode dimensions, as well as for their ideal positions in a smart garment for ECG measurements.

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