

# 50/60 Hz Power Grid Noise as a Skin Contact Measure of Textile ECG Electrodes

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**Abstract:** The electrocardiogram (ECG) is one of the most commonly measured biosignals. In particular, textile electrodes allow for the measuring of long-term ECG without skin irritation or other discomforts for the patient. Such textile electrodes, however, usually suffer from insufficient or unreliable skin contact. Thus, developing textile electrodes is impeded by the often-complicated differentiation between signal artifacts due to moving and breathing and artifacts related to unreliable skin contact. Here, we suggest a simple method of using 50/60 Hz power grid noise to evaluate the skin contact of different textile electrodes in comparison with commercial glued electrodes. We use this method to show the drying of wetted skin under an embroidered electrode as well as sweating of the originally dry skin under a coated electrode with high water vapor resistance.

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



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## 1. Introduction

Cardiac diseases are among the most frequent causes of death in the US [1] and many other countries worldwide [2]. Measuring the electrocardiogram (ECG) of a person is thus one of the procedures often applied in patients with heart-related health issues.

The 12-lead ECG contains six precordial leads according to Wilson, three limb leads according to Einthoven, and three augmented limb leads according to Goldberger and allows full observation of all electrical processes in the heart because of the large number of different measurement directions across the heart [3,4]. For long-term observations, common glued gel electrodes can cause skin irritation and reduce the comfort of the patient. This is why several research projects have aimed at developing textile ECG electrodes [5–10].

The main problem with such textile ECG electrodes is related to skin contact. While commercial gel electrodes create contact between uneven skin and a small metal electrode via a conductive gel, which is soft enough to follow the skin structure, more rigid materials necessitate a certain pressure to enable sufficient contact [11–15]. Moreover, the structure of the textile fabric significantly influences the signal quality [16–19]. In addition, conductive coatings can not only improve skin contact but also result in a thin sweat film, which also improves electric contact [20–23].

In addition, noise from movements, breathing, muscle tonus, and other body-based signals or from the typical power supply of the 50/60 Hz interference has to be canceled, such as by wavelet transform [24–26], adaptive filters [25,27,28], and many others [29–32].

Currently, many approaches to measuring biosignals are based on single-circuit boards, such as Arduino or Raspberry, or even smaller solutions to enable full textile integration [33,34], which may be insufficient for highly sophisticated, real-time filtering methods.

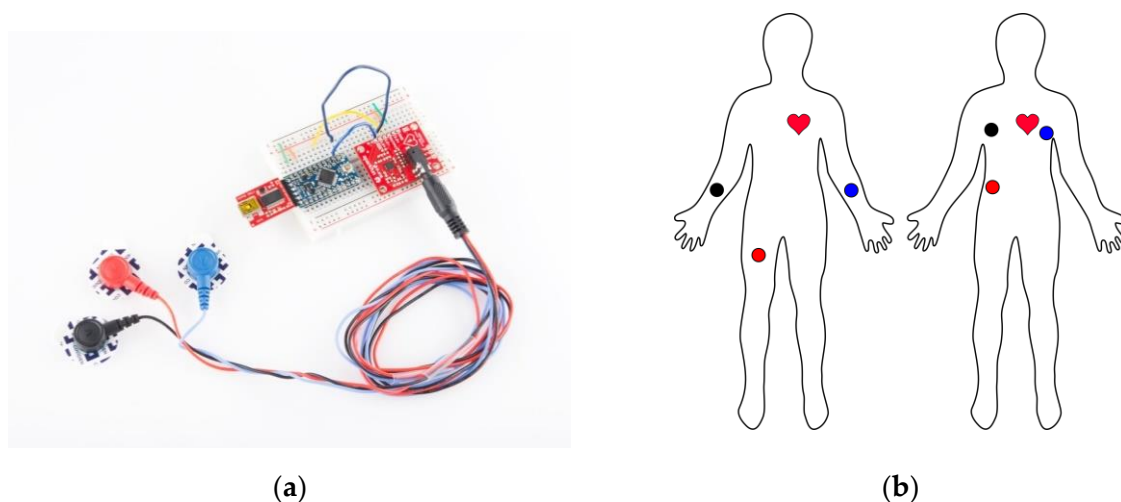
Here, we report on an investigation of textile ECG electrodes with an inexpensive sensor module and an Arduino Uno and without additional filtering. Instead, the 50 Hz noise from the local power grid was used to evaluate the signal quality. Open-circuit noise, which is known to occur in open circuits [35], can, in some cases, be used to investigate

the circuit itself [36]. Here, insufficient skin contact has a similar effect. It is known that 50 Hz noise occurs in ECG measurements because of magnetically induced interference, interference currents in the body, and interference currents in the electrode leads [37], and the real part of the impedance between the skin and the electrode generates so-called Johnson noise [38]. However, applying the 50/60 Hz noise as a qualitative measure of the skin–electrode contact has not been reported in the literature yet.

Our measurements show that this simple approach is highly suitable for unambiguously investigating the skin contact of textile and other electrodes and can thus be applied for electrode optimization in future studies before the filtered signals are depicted.

## 2. Materials and Methods

Measurements were performed using an AD8232 (Analog Devices) ECG sensor module, available from Sparkfun [39] and others. This chip is often used for the research of ECG measurements with textile electrodes [40–43]. The instrumentation amplifier typically has a common-mode rejection ratio from DC to 60 Hz of 86 dB [44]. An Arduino Uno, based on the ATmega328P, was used to read and display the ECG real-time signals on the serial plotter and the monitor of the Arduino IDE, respectively, using the sketch suggested by Sparkfun [45]. The three electrodes were also placed on the chest of a proband as suggested in [39], according to Einthoven. Precisely, one electrode was placed above the left breast, the second electrode was placed above the right breast, and the third one was placed on the lowest rib on the right side of the body. Figure 1a depicts a possible connection of the AD8232 (larger red board) to an Arduino (shown here as an Arduino Nano) and three electrode clips with commercial glued electrodes, while Figure 1b shows possible electrode positions on a human, of which the right one was chosen in this study [39].

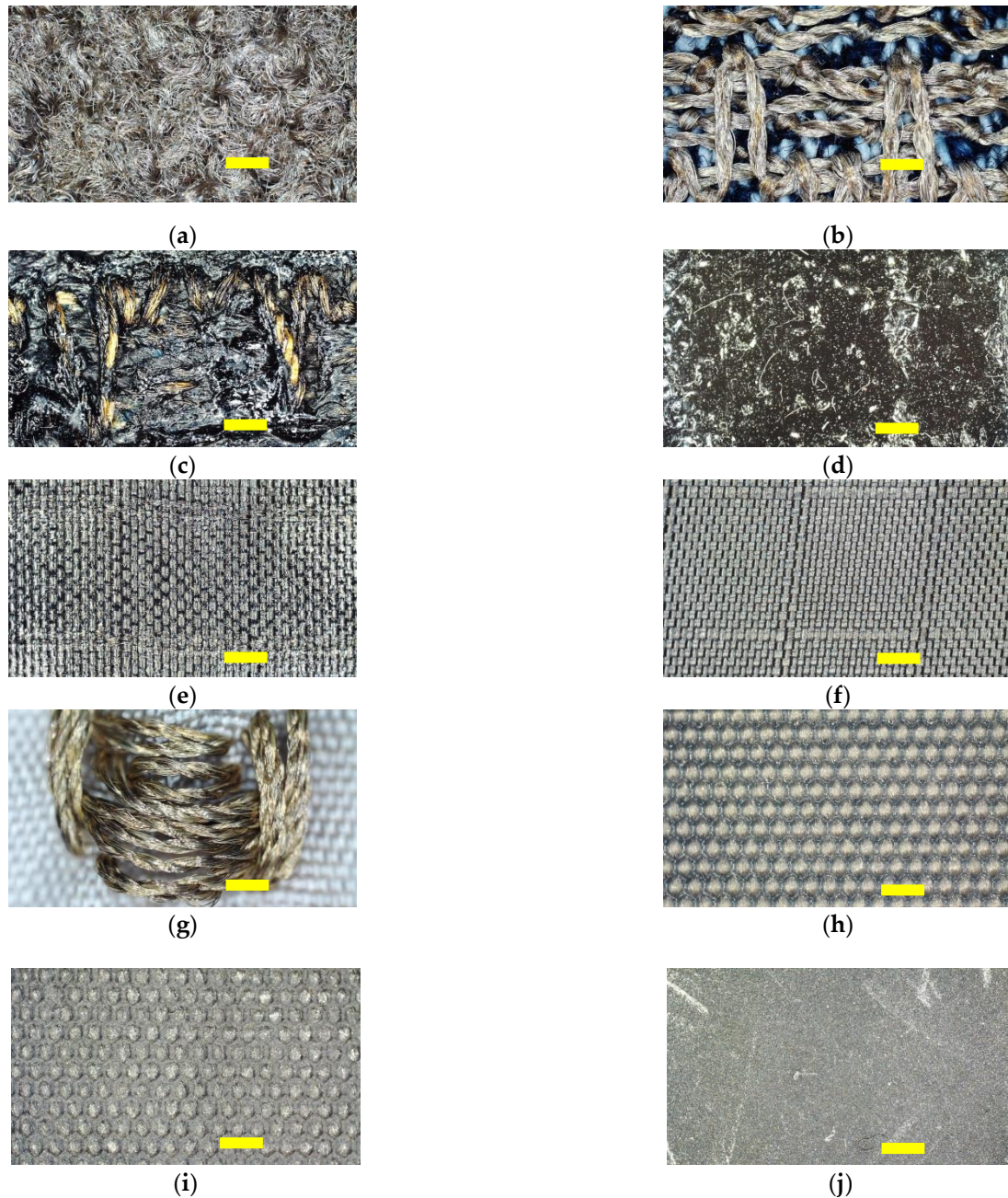


**Figure 1.** (a) Connection of ECG electrodes to the AD8232, which is connected to an Arduino board on the other side; (b) typical electrode positions. From [39], originally published under a CC-BY-SA 4.0 license.

Textile electrodes were produced with areas of approx. 1–2 cm<sup>2</sup>. The electrodes under investigation were as follows (cf. Figure 2):

- Commercial glued gel electrodes;
- Moss-embroidered electrode from silver-coated yarn Shieldex 235/34 dtex 2-ply HC+B, as also used in [46], named “electrode 8” (Figure 2a);
- Electrode with Shieldex backstitch and elastic blind stitch on jeans, named “electrode 4.3” (from [46], Figure 2b);
- Electrode with Shieldex backstitch and elastic blind stitch on jeans, coated by Powersil (20 mg/cm<sup>2</sup>, applied in two layers by a squeegee and hardened for 4 h at 60 °C), named “electrode 4.2” (from [46], Figure 2c);

- Powersil coating on cotton without conductive yarn (Figure 2d);
- Ripstop Silver Fabric (Less EMF, NY, USA; Figure 2e);
- Shieldit Super (Less EMF; Figure 2f);
- Hand-sewn Shieldex yarn on Hansaplast Sensitive Fixation plaster (band-aid, Figure 2g);
- Poly(3,4-ethylenedioxythiophene) polystyrene sulfonate (PEDOT:PSS) double-coated woven fabric (Figure 2h);
- Tubicoat ELH [47] coated woven fabric (Figure 2i);
- Graphite foil, also used in [48] (Figure 2j).



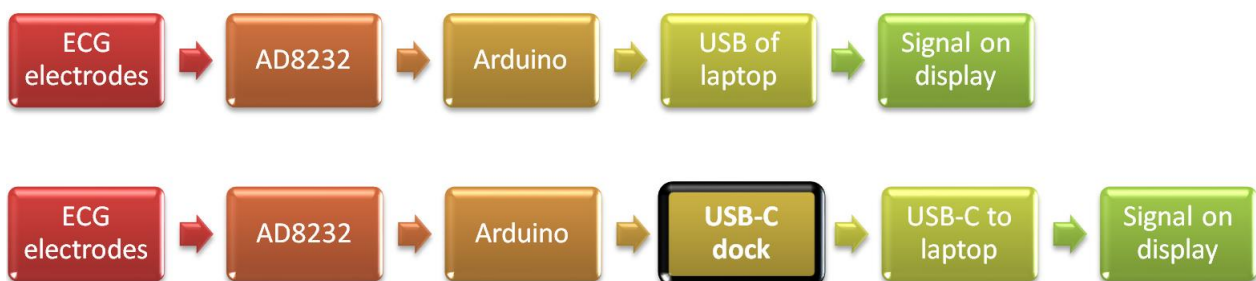
**Figure 2.** Textile electrodes used in this study: (a) electrode 8; (b) electrode 4.3; (c) electrode 4.2; (d) Powersil on cotton; (e) Ripstop Silver; (f) Shieldit Super; (g) sewn Shieldex; (h) PEDOT:PSS coating; (i) Tubicoat coating; (j) non-textile graphite foil. Yellow scale bars depict 1 mm.



The moss-embroidered electrode 8 and the stitched electrodes 4.3 and 4.2 were re-used since they were found to be best suited for bioimpedance measurements [46], suggesting that they may also be suitable for ECG measurements.

All electrodes besides the glued gel electrodes were fixed to the skin using Hansaplast Sensitive. While the electrode–skin contact of commercial glued gel electrodes is typically in the range of 1 k $\Omega$  to 20 k $\Omega$  [37], the skin contact of dry textile electrodes is typically a few hundred k $\Omega$  [49]. It is well-known that increased humidity below the electrode reduces skin contact significantly [4,6]; thus, here, additional tests are shown with wetted electrodes as well as with water vapor-resistant electrodes (e.g., graphite foil and electrode 4.2) under which the proband started sweating after a short time.

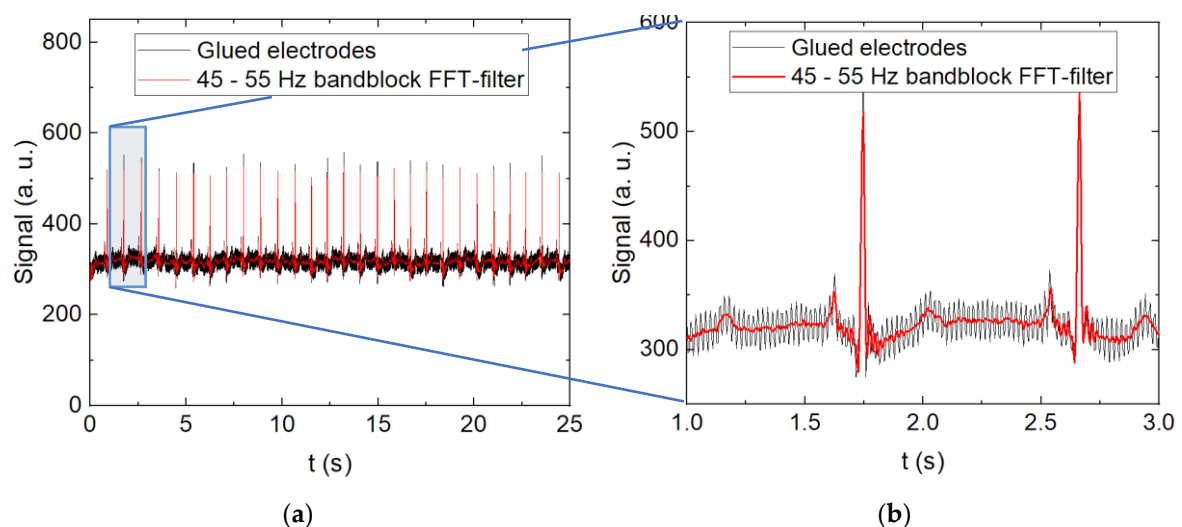
To investigate the impact of the power grid noise, the setups displayed in Figure 3 were chosen. In one setup, a pure laptop without a connection to the power grid was chosen (upper process), while in the other, the laptop was connected to the power grid by a USB-C docking station, over which the signal from the Arduino was also transferred to the laptop. The first setup was thus expected to show nearly no 50 Hz noise, while in the second setup, the 50 Hz noise from the power grid could influence the signal more strongly.



**Figure 3.** ECG measurement with a laptop working in battery mode (**upper process**) or with a laptop connected to the power grid by a common USB-C docking station (**lower process**).

### 3. Results and Discussion

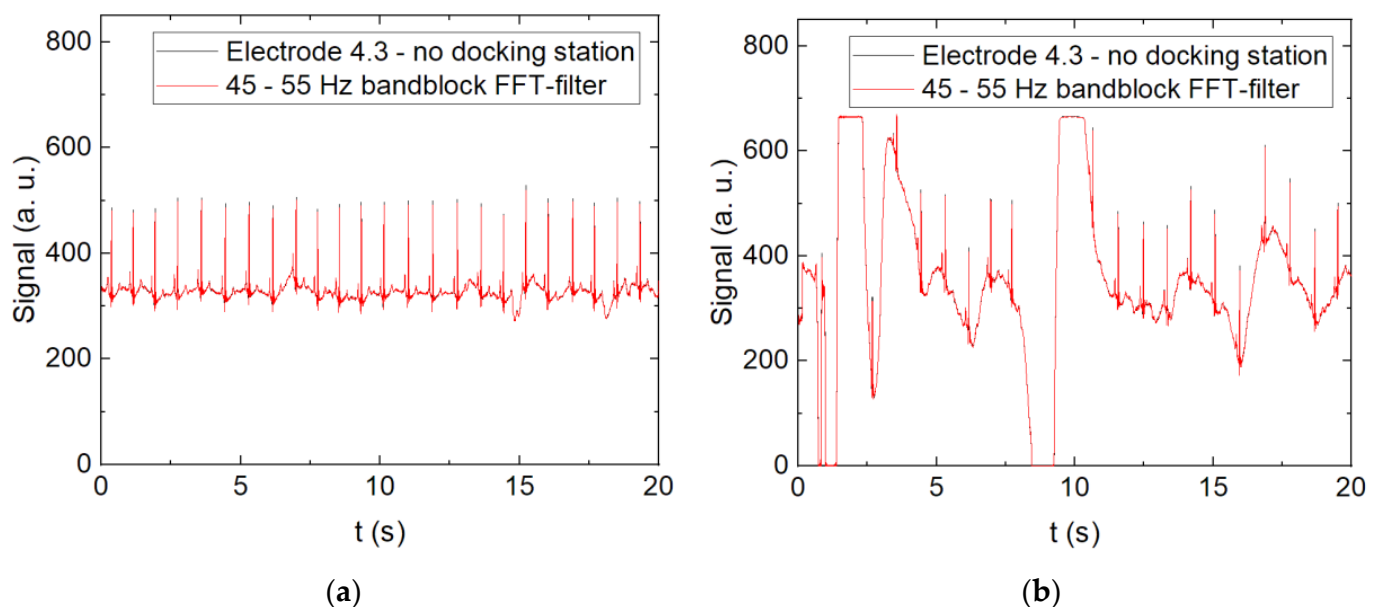
As a reference, Figure 4 shows a measurement with glued electrodes. Data were taken via the serial monitor while the Arduino was attached to a laptop via a docking station, i.e., connected with the grid, in this case working at 50 Hz.



**Figure 4.** ECG measurement with glued electrodes: (a) complete measurement; (b) magnified excerpt of the area marked in (a).

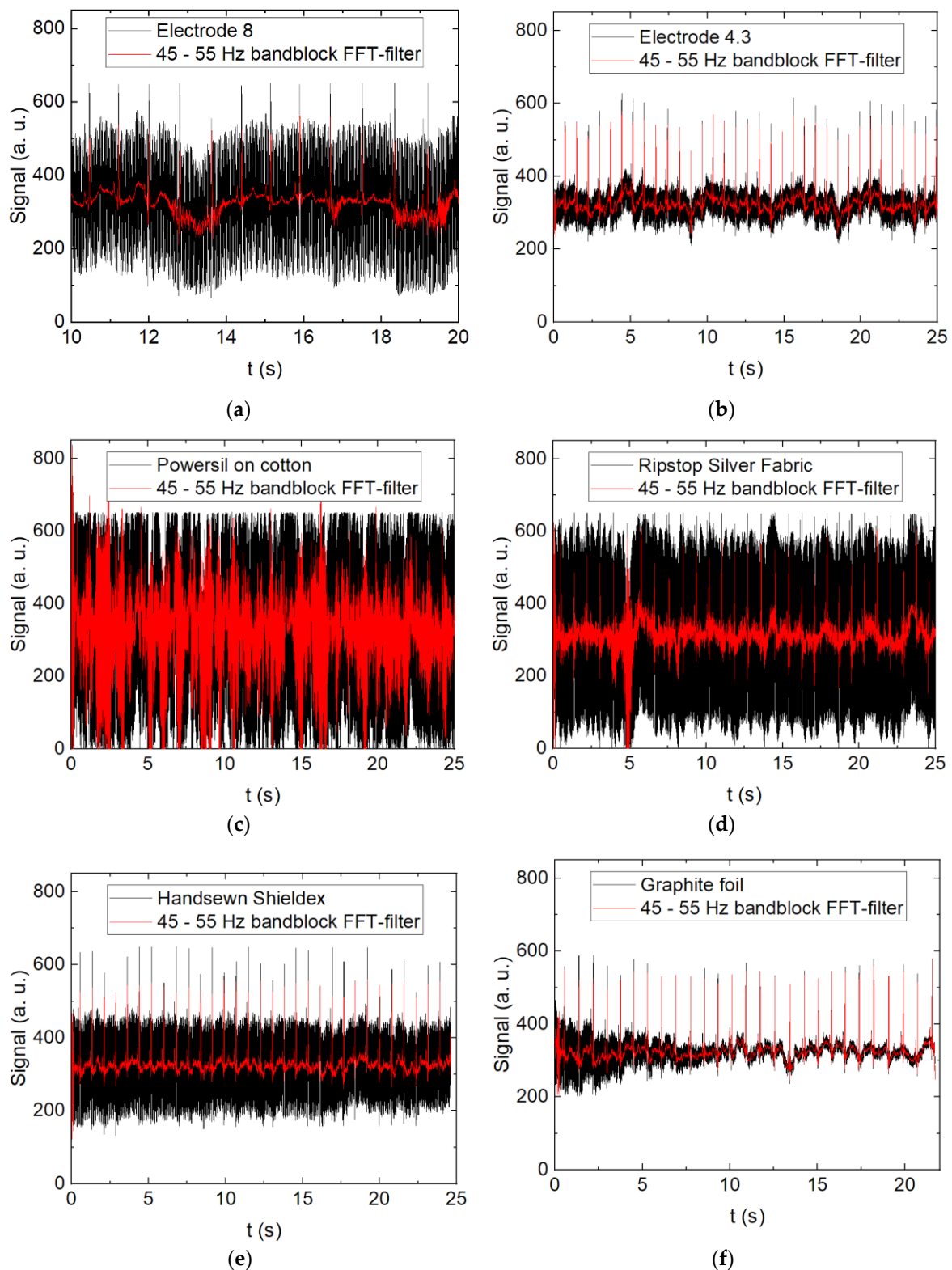
At first glance, Figure 4a shows a noisy measurement with several spikes. Zooming in, Figure 4b reveals that the noise is a 50 Hz interference from the power grid. Apparently, the common-mode rejection—which is used to reduce mains hum—is not sufficient if the Arduino is coupled in the way described here. Further reduction of the 50 Hz noise was performed by a 45–55 Hz bandblock FFT filter applied in Origin 2021 (OriginLab, Northampton, MA, USA). Afterward, a clear ECG signal was visible, with the “spikes” depicting the QRS complexes due to the rapid depolarization of the left and right ventricles (the lower heart chambers), the small P-waves before the QRS complexes showing the depolarization of the atria (the upper heart chambers), and the broader T-waves after the QRS complexes depicting the repolarization of the ventricles.

De-noising the signal using this software filter worked for the complete ECG, as visible in Figure 4a, as well as in all other ECGs that were collected in this study. However, the next tests were performed with the Arduino attached to the pure laptop without a connection to the power grid. The residual 50 Hz noise was very well-filtered by the AD8232 in the case of glued electrodes (not shown here) and for textile electrodes, as depicted in Figure 5.



**Figure 5.** ECG measurement with textile electrode 4.3: (a) sitting without movement; (b) slightly moving.

Figure 5a shows an ECG measurement using textile electrode 4.3 (sewn, without coating) with the proband sitting still. The red (filtered) curve is nearly identical to the black (raw) signal, showing that without direct connection to the power grid, the AD8232 blocked additional 50 Hz noise very well (the same electrode with connection to the docking station is visible in Figure 6b). Nevertheless, a deeper look revealed that the baseline was not flat but showed additional irregular noise. This effect was much more pronounced if the proband moved slightly (Figure 5b), resulting partly in the saturation of the signal and generally in a highly irregular baseline, which makes evaluation of the complete ECG signal quite complicated and nearly impossible as long as no additional high-pass filter suppresses these baseline fluctuations.



**Figure 6.** ECG measurement with textile electrodes with the laptop attached to the power grid: (a) electrode 8; (b) electrode 4.3; (c) Powersil on cotton; (d) Ripstop Silver Fabric; (e) handsewn Shieldex yarn; (f) non-textile graphite foil.

Comparing Figure 5a,b shows the difficulty in evaluating the quality of a textile electrode—while Figure 5a looks nearly sufficient besides some outliers, Figure 5b shows significant problems with the electrodes. It must be mentioned that only quite small

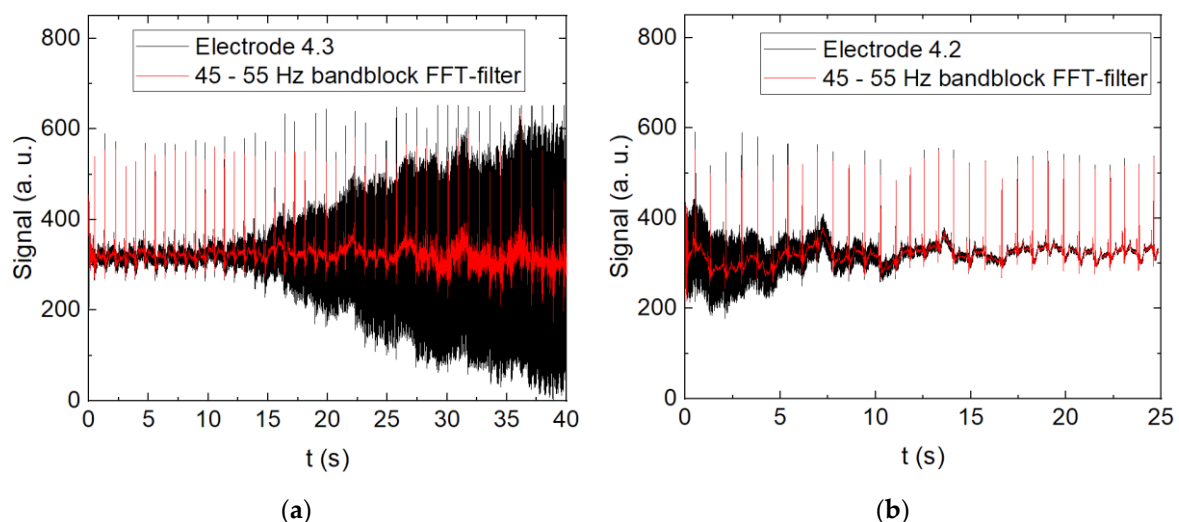
movements were performed during this measurement, i.e., bending the back more or less and sitting up straight or not, without large movements of the arms for which such disturbed signals could be expected even in the case of glued electrodes.

This finding suggests using the normally undesired 50 Hz noise as a measure of the skin–electrode contact. It is well known that 50 Hz noise occurs especially because of poor contact between skin and electrodes [13]. This is also the case for the baseline drift; however, since the latter is also strongly influenced by breathing and moving, the baseline drift is not a reliable measure of the skin contact with the electrodes under investigation. Figure 6 depicts several measurements over 20–25 s taken with the Arduino attached to the laptop via a docking station, i.e., connected to the grid.

In all cases, only the upper left electrode (left arm) was exchanged for a textile electrode, while both other electrodes were glued electrodes to avoid changes in the position as much as possible. The textile electrode was slightly attached to the skin without applying much pressure to reach full skin contact when the tape buckled slightly.

Comparing these measurements, electrode 8 showed high noise with the QRS complexes visible even in the raw signal and T- and P-waves partly visible in the filtered signal. Electrode 4.3 performed much better but still showed a noisy baseline after filtering. Powersil on cotton did not show any signal, not even the pulse (QRS complex). Similarly, no signals could be detected by Shieldit Super, PEDOT:PSS-coated, and Tubicoat-coated fabrics (not shown here). Ripstop Silver performed slightly better but still showed very high 50 Hz noise, and only the QRS complexes were visible after de-noising, while none of the other features could be extracted from the signal. Interestingly, the simple hand-sewn Shieldex electrode also showed a better signal, but it was still worse than the machine-sewn sample 4.3, most likely since the latter has a denser and more fixed yarn distribution (cf. Figure 2). Finally, the graphite foil, a non-textile reference, showed interesting behavior, starting with an average noise that was clearly reduced after some measurement time.

The latter can be explained by the proband's skin starting to sweat below the air-impermeable foil. As mentioned before, this trick is often used to prepare well-working textile ECG electrodes. In Figure 7, the influence of humidity on the skin is clearly visible. Here, no pressure was exerted onto the textile electrodes so that only very narrow skin contact occurred.



**Figure 7.** ECG measurement with textile electrodes: (a) electrode 4.3 on pre-wetted skin during drying; (b) electrode 4.2.

In the case of the sewn electrode 4.3, the skin was slightly wetted before the measurement by putting a few drops of tap water using the fingers at the respective position. Figure 7a shows that the 50 Hz noise started increasing after approx. 10 s when the skin

dried again by evaporation through the electrode. This process was also visible—on different time scales—for all other textile electrodes without a water-vapor blocking coating, such as Powersil.

On the other hand, electrode 4.2, which was Powersil-coated after sewing, showed the opposite time dependence. Here, the 50 Hz noise was clearly reduced after approx. 20 s since the skin started sweating slightly under this air-impermeable electrode. In both cases, the noise of the textile electrodes on slightly wet skin was comparable to the noise of the commercial glued gel electrodes.

As these examples show, the undesired 50 Hz noise can be used as a simple tool to evaluate the skin contact of textile electrodes.

#### 4. Conclusions

Several textile ECG electrodes were evaluated with respect to their skin contact compared with commercial glued gel electrodes. Measurements were performed with an inexpensive ECG module based on the AD8232 chip. The inevitable 50 Hz noise, visible in all measurements in which the laptop for data acquisition was connected to the power grid, was used to evaluate the skin–electrode contact. This simple method is suggested as a possibility to rate the skin contact of textile electrodes in future developments.

Regarding the choice of textile ECG electrodes, none of the tested ones worked well on dry skin. Electrode 4.2, containing a Powersil coating under which the proband started sweating slightly, gave the best results. This indicates that, if the electrodes are not meant for sports but for daily use, a coating that supports sweating is indispensable for textile ECG electrodes.

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