Properties of different types of dry electrodes for wearable smart monitoring devices

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Abstract: Wearable smart monitors (WSMs) applied for the estimation of electrophysiological signals are of utmost interest for a non-stressed life. WSM which records heart muscle activities could signalize timely a life-threatening event. The heart muscle activities are typically recorded across the heart at the surface of the body; hence, a WSM monitor requires high-quality surface electrodes. The electrodes used in the clinical settings [i.e. silver/silver chloride (Ag/AgCl) with the gel] are not practical for the daily out of clinic usage. A practical WSM requires the application of a dry electrode with stable and reproducible electrical characteristics. We compared the characteristics of six types of dry electrodes and one gelled electrode during short-term recordings sessions (≈30 s) in real-life conditions: Orbital, monolithic polymer plated with Ag/AgCl, and five rectangular shaped 10 × 6 × 2 mm electrodes (Orbital, Ag electrode, Ag/AgCl electrode, gold electrode and stainless-steel AISI304). The results of a well-controlled analysis which considered motion artifacts, line noise and junction potentials suggest that among the dry electrodes Ag/AgCl performs the best. The Ag/AgCl electrode is in average three times better compared with the stainless-steel electrode often used in WSMs.

Keywords: dry electrodes; ECG; motion artifacts; signal quality; transient impedance; wearable monitor.

Abbreviations: Ag/AgCl, silver/silver chloride; ECG, electrocardiogram; MG, electromyography; OM, output measure; PCB, printed circuit board; RMS, root mean square; WSM, wearable smart monitor.

Introduction

Wearable smart monitors (WSMs) are becoming practical tools for the improved quality of life because of the development of information and communication technologies (miniaturization and online communication via smartphones or similar devices). The WSM of electrical potentials originating from the heart muscle could function as an online predictor of the need for an urgent intervention.

We show (Figure 1) examples of WSM used for the short-term recordings (up to 30 s) of the electrocardiogram (ECG) that can be sent in the digital form to a remote center for inspection. Many ECG WSMs use stainless-steel electrodes touched by the fingers or palms of opposite arms (Lead I), but other types of dry electrodes made of different composite materials are emerging [1]. The commercially available ECG WSMs for short-term recordings measure the heart rate and, in some cases, detect the atrial fibrillations [2] from Lead I (records between the opposite hands). The new generation of ECG WSMs will allow automatic detection of much more, including the cardiac ischemia (ST segment shift) [3]. In this case, the American National Standards Institute (ANSI) standard for high-pass filtering (HPF) must be followed (cutoff at 0.05 Hz), and raw signal quality is of utmost importance for obtaining high sensitivity and specificity of ischemia detection [4].

The WSM must measure the minimally distorted, noise-free electrophysiological signals, especially if the diagnostics are automatic without expert supervision. The critical points are the electrodes which transform the ionic to electron currents. The noise contaminating the recordings comes from power lines (50 or 60 Hz), motion artifacts due to movements or breathing, muscle electrical activities [electromyography (EMG)], short-term or long-term drifts due to temperature stabilization, the influence of light or changes in the electrode-skin impedances. To minimize the effects, a conductor covered with the hypoallergenic gel is being used. The gelled silver/silver chloride...
(Ag/AgCl) electrodes are considered as the gold standard. The chemical stability of the skin to Ag/AgCl with gel interface minimizes most of the above-listed artifacts and allows the digital filtering to successfully "clean" the recordings [5, 6]. However, the use of Ag/AgCl with gel is not practical, and most of the smart wearable devices tend to integrate reusable dry electrodes.

If dry electrodes are used, then the incidence of motion artifacts is much higher compared with the use of wet (pre-gelled) electrodes [4, 7]. Also, dry electrode half-cell potentials on the electrode-skin junction and junction potentials between the surface layers of the skin are time variable [8].

The current development of an ECG WSM required the analysis of the optimal dry electrodes. We focused on finding the optimal reusable dry electrodes that can be part of the printed circuit board (PCB) with minimal motion artifacts. In this paper, we present the performances of custom-made electrodes made of various types of metals that were mechanically compliant to our design such as stainless steel, Ag, Ag/AgCl and gold. We did not consider rubber-based electrodes because they cannot be integrated into the PCB design. We compared the performances of the commercially available wet electrodes (Top Trace 51 × 33 mm, Ceracarta, Forli, Italy) and Orbital electrodes (Orbital Research Inc., Cleveland, OH, USA) [9]. The output measures (OMs) were related to noise and motion artifacts in ECG-free signals recorded from the fingers as we investigated the electrical characteristics of the electrode-skin interface.

The characterization of electrodes in stationary conditions or long-term records is well documented in the literature [10, 11], but the application of WSM held in hand does not belong to the stationary conditions. Therefore, we evaluated artifacts during the transitional changes in the electrode-skin interface (=30 s from touching the electrodes). The frequency spectrum of ECG signals overlapped with artifacts; therefore, it is impossible to clearly separate ECG from the artifacts. If we record ECG-free signals with the same electrodes positioned at the same skin areas, then we measure only artifacts. These signals would simply superpose on the ECG action potentials when the ECG is measured. That is why we positioned the two electrodes at two fingers of the same hand to minimize the effects of electrophysiological signals (muscle activities and ECG). A similar method was suggested by Searle and Kirkup [10] who compared motion artifacts from three different pairs of electrodes by attaching them rigidly to the same housing with a fixed frequency vibrating element and placing them on the surface of one forearm. Our approach introduces three fundamental differences compared with what has been done: (1) the position of the electrodes was on the actual place where they are used for recording in ECG WSM (on the fingertips), (2) motion artifact was naturally induced in two typical positions of the forearm and (3) multiple OMs were calculated from the statistical analysis of the repeated recordings.

Materials and methods

Protocol and instrumentation

Instrumentation: Custom-designed holders for fingers (Figure 2, lower panel) supported seven sets of electrodes made of different materials and shapes (Figure 2, upper panel):

1. WET: a commercial wet electrode, Top Trace from Ceracarta (51 × 33 mm, metal contact diameter: 7 mm, hydrogel diameter: 17 mm)
2. ORB: a monolithic polymer plated with Ag/AgCl from Orbital (diameter: 25 mm, effective surface: 500 mm² and pin height: 150 µm, front to back resistance: 0.4 Ω)
3. ORB CUT: we cut the Orbital electrode in the rectangular shape (10 × 6 × 2 mm) to fit our device design (front to back resistance: 8 Ω)
4. AG: a custom-made Ag electrode in the same shape as ORB CUT (front to back resistance: 0.5 Ω)
5. AG/AGCL: a custom-made Ag/AgCl electrode in the same shape as ORB CUT (front to back resistance: 0.4 Ω)
6. AU: a custom-made gold electrode in the same shape as ORB CUT (front to back resistance: 0.5 Ω)
7. INOX: a custom-made stainless-steel (AISI304) electrode in the same shape as ORB CUT (front to back resistance: 0.5 Ω).

We chose all metal electrodes manufactured in the shape of Orbital electrode because they can penetrate the hair on the chests in future use. The preparation of the custom-made electrodes was done in the institution workshop to avoid any edge effects. We present the results of the evaluation of the electrode-skin impedance for each type of electrode in the Appendix.

A host computer operating in the Windows environment captured the signals. The digital amplifier, which integrates a 24-bit analog-to-digital (A/D) converter (ADS1298, Texas Instruments, Dallas, TX, USA) with the gain of the input instrumentation amplifiers set at $A = 12$, in the direct current (DC) coupled mode was connected to the electrodes.

Subjects: Ten volunteers [four men and six women, age 35 ± 11 years, body mass index (BMI) 23.9 ± 4.3 kg/m²] with no known skin, adipose tissue, vascular disorder or pathological tremor volunteered in the study. All subjects signed the informed
consent approved by the Ethics Committee of the Medical School, University of Belgrade, Belgrade.

Protocol: The subjects were sitting comfortably next to a table, with one elbow supported by the table. The measurements included two setups (Figure 2, lower panel): (1) the forearm was resting on the table, and the subject was holding the electrodes on the sides of the device with his/her thumb and middle finger, and (2) the forearm was pointing against the gravity. Subjects were instructed to relax as much as possible (a minimum contraction of forearm muscles and no motion). Three sessions, lasting 30 s each, were recorded for both setups. In total, 420 sets of signals were recorded (6 per subject × 7 electrode types × 10 subjects), or 6 × 10 × 30 = 1800 s of recordings in natural conditions per each electrode type.

Data analysis

All data processing was done in Matlab 2014b (MathWorks, Natick, MA, USA). We used six recordings per electrode type per subject to estimate the mean ± standard deviation (SD) values for each OM. All records were heuristically inspected for unlikely irregularities (e.g. excess 50 Hz noise due to poor contact, recording interrupted or subject moved during recording). If the inspection suggested high contamination by the noise of non-physiological source other than 50 Hz, then these signals were excluded from further analysis.

As there was a substantial difference in absolute values between subjects, we normalized values for all electrodes in each subject with the following formula:

\[
\text{OM}_{i}^{\text{Normal}} = \frac{\text{OM}_i}{\sum_{j} \text{OM}_j}
\]

where OM is the output measure, and i and j represent the type of electrode.

Seven different OMs were developed to quantify various aspects of noise and artifacts in ECG recordings and their influence on proper ECG signal reading and diagnosis. Signal processing steps and OMs were the following:

1. OM 1: The root mean square (RMS) value of 50 Hz power-line noise (a band-pass filtered original signal with third-order Butterworth filter between 48 and 52 Hz).
2. OM 2: The RMS value of 6–13 Hz tremor (a band-pass filtered original signal with third-order Butterworth filter between 6 and 13 Hz). Physiological tremor is expected oscillatory motion in fingers of all healthy individuals in the frequency range of 6–13 Hz with different amplitudes depending on the current physiological and psychological conditions [12, 13].
3. The component of the signal from the muscle activities (EMG): After subtracting 50 Hz noise and tremor (the same filtered signals which were used to calculate OM 1 and OM 2) from the original signals, the signals were low-pass filtered with third-order Butterworth filter at 30 Hz to remove EMG signals. The EMG frequency range is 30–300 Hz. The EMG component can be seen in Figure 3A.
4. The HPF signals at: (a) 0.05 Hz [finite impulse response (FIR) filter with order 4496] or (b) 1 Hz (third-order Butterworth filter). We selected the cutoff frequency at 0.05 Hz based on standards for devices used for the detection of ischemia (ST shift) and other heart pathologies. The 1 Hz cutoff was chosen for the detection of heart rhythm (occurrence of QRS segments).
5. OM 3: The RMS values of the signals after the HPF. These values provide a measure of total baseline wandering during 30-s recording (there was no ECG in the recorded signals). Large baseline wandering can make ECG signal reading difficult if there is no auto-scaling function on the reading device.
6. OM 4: The histograms with seven BINs after each type of HPF:
   (1) BINO01: number of samples with values ≤0.01 mV
   (2) BINO05: number of samples with values between 0.01 mV and 0.05 mV
   (3) BINO1: number of samples with values between 0.05 mV and 0.1 mV
   (4) BINO2: number of samples with values between 0.1 mV and 0.2 mV
   (5) BINO5: number of samples with values between 0.2 mV and 0.5 mV
   (6) BIN1: number of samples with values between 0.5 mV and 1 mV
   (7) BINout: number of samples with values >1 mV (typical amplitude of QRS segment recorded on the surface of the skin)
7. The duration of intervals in which all consecutive samples are smaller than 0.1 mV after each type of HPF. This threshold was selected because it represents a minimal ST shift that can be interpreted as pathological [14]. This segment was named a “good” signal interval. OM 5: We calculated the longest “good” interval.
8. OM 6: The number of “good” segments lasting at least 5 s for each type of HPF. The duration of 5 s was selected because in the most common case there would be at least five heartbeats within 5 s. If the signal is artifact free during this period, it should be enough for a cardiologist to notice if there are any essential changes in the shape (not rate) of the ECG signal.
9. OM 7: The number of “good” segments lasting at least 2 s after each type of HPF. The duration of 2 s was selected because in general case there would be at least one heartbeat within 2 s. Considering that the ECG signal can be randomly contaminated with noise and artifacts, if there were multiple 2-s “good” intervals, then they can be selected to form a sequence of good-quality heartbeat signals for the proper diagnosis of potential ischemia by the cardiologist.

Examples of how we calculated the number of “good” segments are as follows: if a good segment (all values <0.1 mV) lasts 3 s, then it contains one 2-s “good” segment and none of 5-s “good” segments. If a “good” segment lasts 6 s, then it includes three 2-s “good” segments and one 5-s “good” segment. If there are two “good” segments lasting 2 s and 6 s, then the recording contains four 2-s “good” segments and one 5-s “good” segment.

Statistical analysis was done in an Excel Xlstat program (Microsoft Office). Data were tested for normality using the Shapiro-Wilk normality test. Due to the relatively small sample size in the experiments, we used non-parametric tests. The Kruskal-Wallis test tested the significance of the results for independent samples with the significance level \( p = 0.05 \).

Results and discussion

The graphical presentation of all calculated OMs is shown in Figure 3. The heuristic examination of signals resulted
in the exclusion due to biological noise of just over 5% of the recorded signals (22 out of 420 recordings). Different RMS values for all subjects and all electrodes are shown in Figure 4. Summary statistics for each electrode type and each OM are shown in Figure 5. The normalized values for all electrodes in each subject (e.g. in Figure 4, the RMS of 50 Hz noise for subject 4, on AU and WET electrodes, is 0.47 ± 0.25 mV and 0.03 ± 0.01 mV, which equals to 33.9 ± 18.3% and 1.9 ± 0.9% when normalized for subject 4, respectively) were used to calculate mean ± SD values for all subjects for each electrode type (Figure 5).

The Orbital electrode in original form (ORB) and AG picked up the least of 50 Hz noise compared to other dry electrodes. Cutting Orbital (ORB) to a rectangular shape resulted in 45% more noise, which was expected as decreasing the contact surface per se increases the impedance, which lowers the signal-to-noise ratio. Also, by cutting the surface conductive layer, we may have deteriorated the characteristics of the applied technology. The AG/AGCL electrode picked up 5% less noise than ORB CUT. There was no statistically significant difference between the AG/AGCL, ORB, ORB CUT and AG electrodes. AU and INOX performed significantly worse than other tested electrodes. WET, AG/AGCL and ORB induce similar RMS values of tremorous finger motion between 6 and 13 Hz, while ORB CUT and AG recorded slightly larger amplitudes. AU and INOX performed significantly worse.

After HPF at 0.05 Hz, the RMS value of the baseline wandering was minimal for ORB (7.04%), followed by AG/AGCL (8.29%), ORB CUT (8.54%), AG (10.77%), AU (20.59%) and INOX (41.13%) compared to WET (3.64%). After HPF at 1 Hz, the RMS value of the baseline wandering was minimal for AG/AGCL (7.45%), followed by ORB (8.31%), ORB CUT (10.54%), AG (10.84%), AU (20.59%)
and INOX (37.01%) compared to WET (5.25%). AU and INOX were statistically worse than other electrodes in both cases of filtering.

After HPF at 0.05 Hz (clinical standard), ORB, ORB CUT and AG/AGCL had a similar number of samples in all BINs. ORB had the highest number of samples in the first three BINs (values up to 0.1 mV) (70.88%) compared to WET (82.97%), AG/AGCL (67.93%) and ORB CUT (65.93%) followed. After HPF at 1 Hz (dynamic filtering), AG/AGCL had the highest number of samples in the first BIN (values up to 0.01 mV) following the WET closely. However, the number of samples in the first BIN is irrelevant in the applications in which 1-Hz filtering is justified.

Histograms and quantity of low-value BINs are essential for distinguishing if the electrodes are more or less prone to noise and baseline wandering. Histograms are not good indicators of unpredicted abrupt changes in signals. For this purpose, we developed a new measure – the duration

![Figure 4: Absolute RMS values of different signal components originating from: power-line noise between 48 and 52 Hz, motion artifact due to physiological tremor between 6 and 13 Hz, baseline wandering after removing 50 Hz noise and tremor and band-pass filtering between 0.05 and 30 Hz or 1 and 30 Hz. Standard deviations show the variability of results from six recordings in each subject. Numbers next to individual bars represent values of the bar that is out of the presented range.](image-url)
Figure 5: Summary statistics for different output measures for seven electrode types. Top panels show a comparison of normalized RMS values of 50 Hz noise and tremor motion artifact. After low-pass filtering (LPF) at 30 Hz and high-pass filtering (HPF) at 0.05 Hz (gray side) or 1 Hz (red side), RMS values quantify baseline wandering. Stacked bar plots represent values from the histograms divided into seven custom BINs. The length of the “good” segment (all samples in segment <0.1 mV) is shown relative to the length of the whole recording (the 30 s). The number of “good” segments lasting at least 5 s or 2 s is shown in bottom panels as orange and blue bars, respectively.
of “good” parts of a signal (where all the consecutive values are smaller than 0.1 mV). After HPF at 0.05 Hz, the longest “good” interval in the 30-s recording was obtained with AG/AGCL (9.08 ± 5.67 s) compared with WET (15.78 ± 10.07 s). AG (796 ± 6.82 s) and ORB CUT (795 ± 3.39 s) had better behavior than ORB (6.99 ± 4.63 s). AU and INOX barely had any 2-s “good” intervals. After HPF at 1 Hz, the average duration of the longest “good” part of the signal was the best for AG/AGCL (24.6 ± 4.06 s) compared with WET (26.74 ± 5.78 s). AG had the highest number of “good” segments lasting at least 5 s each (4.75 ± 2.41 s), but the results for WET, AG/AGCL, ORB, ORB CUT and AG were comparable. AU and INOX performed worse.

The results of the signal quality analyses are not at all in line with the results of electrode-skin impedances presented in the Appendix. An example is that the AG/AGCL electrode had high values of impedance, but still recorded well the signals from the skin surface. This statement is in line with Chi et al. [1] presenting that low resistance (high conductance) is not essential for good electrode performance, and that maximizing resistance (minimizing conductance) in electrode-skin coupling could be beneficial in specific cases.

The results in favor of Ag/AgCl-based electrodes are following our previous work (unpublished results). We tested ORB CUT electrode performance in a study that included 2096 recordings from 34 subjects using prototype hand-held devices in the home environment. We found that the signal quality was acceptable for the detection of ST shift in 95% of the recordings.

**Conclusion**

We compared the performance of different dry, eventually PCB mountable, non-expensive materials that could be used as dry electrodes in ECG WSM. The gold standard is the Ag/AgCl electrode with gel, described in the literature [15]. We measured the signals coming from the body at the skin with different dry electrodes in real-life situations and classified them into seven metrics. The results suggest that Ag/AgCl-based electrodes perform best regardless of the shape and large inter-subject variability. ORB minimizes the power-line noise, introduces small baseline wandering and allows the most extended good-quality recording sections when compared to other dry electrodes. The next in line are the Ag/AgCl dry electrodes picking up little more power-line noise and comprising a most substantial number of short sequences of good-quality signals. Electrodes based on other materials showed worse characteristics in our measurements. INOX performed the worse in all metrics.

Filtering at 1 Hz substantially lowers the baseline wandering and other motion artifacts, leading to less differentiation in the behavior of different dry electrodes.

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**Figure 6:** Experiment setup for recording of electrode-skin impedance and model of the electrode-skin interface (modified from [1]). $E_{hc}$ is a half-cell potential due to interaction of the skin humidity and sweat with the electrode.
compared to each other and compared to wet electrodes. That means that in applications such as detection of heart rate or atrial fibrillations with HPF at 1 Hz the use of INOX electrodes may be justified as an acceptable trade-off between the price, durability and quality. However, for the applications in ECG WSM with algorithms for the automatic detection of cardiac ischemia and other heart diseases based on the assessment of the morphology of the ECG signals, the ANSI standard for HPF must be followed (HPF limit at 0.05 Hz). In those applications, much better performance can be achieved with Ag/AgCl-based electrodes.

We used seven metrics to evaluate the characteristics of dry electrodes made of different materials. Depending on the application, different metrics can be more important. For instance, if the electrode is used in a WSM device to automatically detect ischemia (with HPF at 0.05 Hz), then the most significant differences are found in the number of 5-s segments with all signals <0.1 mV. If the WSM is used to detect arrhythmias (with HPF at 1 Hz), then the most significant differences are found in RMS values after filtering. The results of our study suggest that the electrode-skin impedance is not necessarily a good indicator of the electrode performance for short-term recordings.

Appendix

Electrode-skin impedance was measured in the 10 subjects described earlier, with the experiment setup shown in Figure 6. OP177 is an operational amplifier (OP) with a high input impedance (45 MΩ). One electrode was connected to the negative input of OP and the other electrode of the same type was connected to the output of the OP. Electrode pairs were positioned on a plexiglass plane. The subject touched with his/her left and right index finger two electrodes of the same type on different planes.

Figure 7: Results of electrode-skin impedances for different electrodes in one subject.
(A) Signals Vin (blue) and Vout (red) from Figure 6. And moment when the skin touches the electrodes (black vertical line); (B) FFT calculated in four time windows [0 s, 2 s], [3 s, 5 s], [15 s, 17 s] and [28 s, 30 s] starting from the moment when skin touches the electrodes; (C) Real and imaginary part of impedance; (D) Absolute values of impedance vs. frequencies in four time windows; (E) Absolute values of impedance vs. time for selected values of frequencies. Presented values are for two electrode-skin contacts in series (two fingers on two different electrodes).
$V_{in}$ is a complex low-voltage periodic signal composed of 22 sine waves at different frequencies, described by the term:

$$V_{in} = \sum_{i=1}^{22} 0.5 \sin (2\pi i t^i)$$

$V_{in}$ was generated as an analog output on NI6363 USB DAQ board, in LabView program (National Instruments, TX, USA). For each subject, $R_{current}$ was selected to one of the values (500 kΩ, 5 MΩ, 50 MΩ) based on the highest values of impedance in the subjects to avoid saturation of $V_{out}$ and optimize the resolution of 16-bit AD conversion. $V_{out}$ was acquired on one analog input by the same LabView program. Each acquisition lasted 40 s. The subject placed the fingers on electrodes after the acquisition started, to ensure the recordings from the instant when the skin touched the electrode (example in Figure 7A). After touching the electrodes, the palms were resting on the plexiglass board to minimize the motion.

From the recorded signals, the program automatically detected the instant when the skin touched the electrodes, as the position of the time window of width $f_s$ ($f_s$ is sampling rate) in which the equation:

$$\max(V_{out}) < 0.5 \times \max(V_{in})$$

was satisfied for the first time (black line in Figure 7A).

We calculated the fast Fourier transform (FFT) of $V_{in}$ and $V_{out}$ in time windows of $2*f_s$ (example in Figure 7B):

$$F_{in}(j) = \text{FFT}(V_{in}[k])$$

$$F_{out}(j) = \text{FFT}(V_{out}[k])$$

Impedance was calculated from the formula:

$$Z = \frac{F_{out}}{R_{current} \times F_{in}}$$

The impedance values for time windows [0, 2 s], [3 s, 5 s], [15 s, 17 s] and [28 s, 30 s] are shown in Figure 7C (real and imaginary parts) and 7D (absolute values). Transitional time behaviors of impedances for selected frequencies ($|Z(j)|$ where $j$ corresponds to frequencies of 1, 3, 8, 16, 25 and 81 Hz, are shown in Figure 7E.

The same trends were found in all subjects. The only differences were the maximum and minimum values of the impedance (Figure 8).

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