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Novel dry electrodes for ECG monitoring

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Abstract
The development, fabrication and characterization of two novel dry bioelectrodes—conductive and capacitive ones—for biopotential monitoring are presented. The new electrodes have the potential to improve the applicability of dry electrodes in ambulant recording of ECG by reducing motion artifacts as well as the contact impedance to the skin. Furthermore, a passive filter network is integrated into the new electrodes to suppress slow offset fluctuation of the ECG signal caused e.g. by motions like breathing or changes in the electrode–skin interface properties. Compared to standard gel electrodes these new electrodes exhibit equivalent and superior contact impedances and biosignals. The integrated filter network effectively suppresses fluctuating offset potentials.

Keywords: biopotential, bioelectrode, ECG, dry electrode, capacitive electrode, artifact

1. Introduction

Rapid advancement in medical technologies in the past has contributed to significant improvements in patient care. Partly because of technological advances, within the last 20 years the life expectancy has shifted from about 72 years to 80 years and still increases. At the same time, the costs for health care have increased due to novel more expensive medical treatment. The challenge for engineers in conjunction with doctors is to develop new or to improve the methods of preventive care and decrease the costs of instrumentation as well as its use and maintenance. Especially, microsystem technologies offer numerous ways to generate miniaturized and innovative medical systems, superior to standard technologies in terms of material costs and reliability. Furthermore, miniaturizing such systems can increase the patient comfort considerably (Karilainen *et al* 2004).

Cardiovascular diseases are the main cause of death within the population in the age range 44–64 years, and the second most frequent cause of death of people between 24 and

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44 years. In Germany, for example, about 300 000 people suffer from a heart attack annually. An early recognition of symptoms, and warning of the patient or doctor, would enable preventive actions to avoid the attack and thus the risk of irreparable damage to organs or even death. Monitoring risk groups, such as people who recently were subject to bypass surgery or pacemaker implantation, can effectively decrease the number of heart attacks. Long-term recording of ECG is a standard procedure in current cardiac medicine, but mostly due to the limited electrode performance of presently available systems monitoring is restricted to a few days only, even though much longer recording times are of clinical interest.

The objective of this work is an engineering, not a clinical approach, to generate a new type of dry ECG electrodes exhibiting reduced contact impedance and motion artifacts, with expected improved long-time stability. Various dry conductive and capacitive electrodes were designed, fabricated and characterized. Furthermore, using a special electrode configuration a method for a passive suppression of slowly changing offset potentials was developed. This new configuration will also allow a batch processing of electrodes including electronics as microsystems.

2. Biomedical basics

Biopotentials are electrical potentials inside a living body, which are created due to distributed ions. Excitable cells of the heart muscle are the origin of ECG. Due to the conductivity of the human body these potentials are carried to the body surface, where they can be measured. Na^+, Ca^+ and Cl^- ions are responsible for the charge transport in an organic system in contrast to electrons in the leads of an ECG device. Consequently, ion currents have to be converted to electron currents with the electrode as the transducer. The skin, which has a dry dielectric outer layer, the stratum corneum, impairs the transfer from ions in the tissue to electrons in the electrode. The capacitance of this layer is poorly defined and unstable, as it depends on skin properties such as humidity and thickness. Typical values can be as low as 100 nF cm^{-2} (Taheri *et al* 1994). In addition to the skin impedance, the electrical transducer comprises the resistance of the electrolytic gel and the double layer at the electrode–electrolyte interface, as well as the half-cell potentials at both electrolyte interfaces. All these contributions can be described by an electrical equivalent circuit shown in figure 1 (Webster 1998).

Differing ECG signals are obtained depending on the electrode location on the body. Since the measured signal is a superposition of all electrical activities in the body–electrode setup, actions of the patient, such as walking, will cause additional signals of extracardiac origin, described in section 3.2. These motion artifacts may be misinterpreted as extra systoles. Fast extracardiac muscle motions may even disturb the ECG to a hardly recognizable condition. Such artifacts can be reduced by using suitable electrode sites. Slow motions such as breathing introduce fluctuations in the zero line. When the lungs fill up with air, the position of the heart changes relative to the electrodes and thus the voltage between them. Furthermore, stretching the skin under the electrode changes the electrical properties of skin and interface, which may introduce additional artifacts. During the analysis of an ECG signal, all these phenomena have to be accommodated to avoid misinterpretations.

The standard efficient transducer is the weakly polarizing Ag/AgCl gel electrode. In contrast to dry electrodes, which are denoted as partly polarizable, they introduce very low ohmic impedances into the equivalent circuit. Dry electrodes can be modeled by a parallel circuit of an ohmic and a capacitive impedance, with an additional Warburg resistance (Geddes and Valentinuzzi 1973). The capacitive electrode is modeled as perfectly polarizable, i.e. it introduces a capacitor C_d.

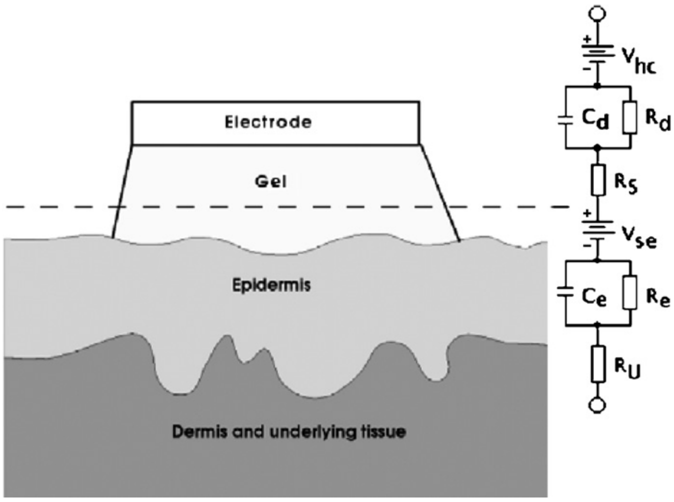

Figure 1. Cross-section of the skin–gel–electrode interface, and its electrical equivalent circuit (Webster 1998).

The task of the moist electrolyte gel in Ag/AgCl gel electrodes is to reduce the impact of the skin on the impedance by making its dry outer layer ion conductive. The standard Ag/AgCl gel electrodes have limited shelf life and are not reusable. Because of dehydration they can be used for a few days only. Dehydration modifies the electrode impedance, which generates noise and other artifacts. Furthermore, the gel can cause skin irritations and support bacterial growth (Searle and Kirkup 2000).

Instead, dry and capacitive electrodes could improve the long-term performance of an ECG setup. Even though some positive results have been published (Searle and Kirkup 2000, Prutchi and Sagi-Dolev 1993), actual dry electrodes, which are rigid metal plates, generate evident problems: in ambulant monitoring, where the patient is able to move freely, the electrodes shift relative to skin. They have less defined and higher contact impedance in coarse hair, locally isolating the electrode from the skin, producing an undefined contact area. Further, especially the capacitive electrodes may charge due to friction because of reduced moisture. To overcome these problems, the skin has e.g. been mechanically punched (Griss *et al* 2001). However, besides the high risk of inflammation even this invasive method is not stable in the long term since the skin regenerates and encapsulates the electrode with scar tissue to isolate it from the body fluids.

In ambulant monitoring, a patient is shaved before applying the electrodes, which is arduous and ineffective for long-term monitoring since hair will regrow fast. Also the mechanical abrasion of the skin, a standard method in ambulant care, does not help since skin regenerates within about 24 h.

During use, however, the skin under a tight electrode will moisturize due to sweat, i.e. the interface impedance will drop. In spite of the lower ion conductivity of sweat as compared to a gel the interface impedance will become low enough for safe signal detection.

3. Novel dry and capacitive electrodes

To overcome most of the above-mentioned problems, a novel soft dry electrode was realized. It adapts to the skin topography such as curvatures and hair, and guarantees small relative

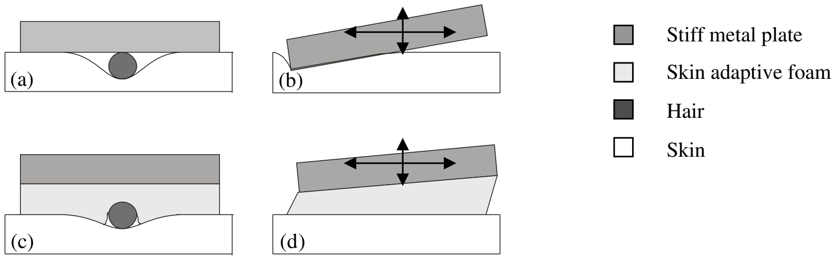

Figure 2. Advantages of foam electrodes in comparison to stiff dry electrodes. (a) Loss of contact area and increase of contact impedance due to hairs. (b) Loss of contact and increase of artifacts due to charging effects caused by motion of electrode relative to skin. (c) Increased contact area. (d) Contact maintained during movements.

motion of the skin to the electrode because of its flexibility and cushioning effect (figure 2). In addition, a novel concept was developed, which suppresses unwanted slow sub-Hertz offset potential fluctuations due to drift in the electrode properties or patient motions like breathing. A passive filter network integrated into that electrode setup acts as a passive high-pass filter directly at the origin of the signal.

3.1. Reduction of contact impedance

The electrode–skin interface impedance is governed by contact area and skin properties. For low impedance the contact area should be large. On hairy skin, air gaps hundreds of micrometers wide can occur between the electrode and the skin, which will significantly reduce the area. Skin adaptive electrode e.g. conductive foam, soft enough to adapt the geometry of the hair, can significantly reduce this effect (figure 2(c)) as long as the hair does not completely conceal the skin, whereas rigid electrodes will fail even at loose occlusion. Under slight pressure, the contact of foam electrodes on hairy skin will be nearly as good as that of rigid electrodes on hairless skin. This will be shown in section 5.1.

3.2. Reduction of motion artifacts

The motion artifacts generated in biopotential electrodes are induced by charging due to friction and slipping, as well as a change in the electrical double layer at the interfaces, e.g. because of deforming and pressing the skin. Since the skin and electrode differ in their electron population and work function, electrons are transferred through the interface. Random separation and contact in the metal–dielectric system generates charge fluctuations since equilibrium in the metal is restored nearly instantly, but slowly in a dielectric, which in turn induces charges in the metal in a separation–contact sequence. A soft electrode will maintain the contact even under motion and also rubbing and sliding of the electrode on the skin will induce no charge. Attaching the electrode with a little pressure, and its elasticity, stabilize the contact both horizontally and vertically (figure 2(d)). Even if movement occurs, only parts of the electrode move and the overall contact area will hardly change.

3.3. Dry electrodes as a passive filter network

As one heart cycle takes about 1 s, and the QRS complex lasting less than 0.1 s generates the highest frequencies, the bandwidth of 1 Hz to 250 Hz is relevant for ECG. Often the recommended cutoff frequency for ECG amplifiers is set lower than 1 Hz, but this is not

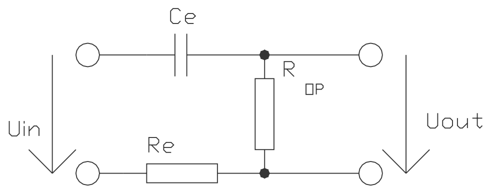

Figure 3. Schematic presentation of a passive filter electrode network comprising ohmic and capacitive electrodes, as well as the input impedance of the operational amplifier.

necessary in monitoring applications, where a high fidelity is not required, as will be seen in our results. Spurious signals, such as the zero-line fluctuation with frequencies below the chosen cutoff frequency, are commonly suppressed by a high-pass filter in the signal amplifier. Since the heights of these fluctuations often surpass the heart signal significantly, the suppression at their origin would be favorable, i.e. the filter should be integrated into the electrodes. A combination of ohmic and capacitive impedances (figure 3) of differently designed electrodes can introduce such a filter function in the transducer as long as the input impedance of the amplifier following the network exceeds the electrode–skin impedance.

4. Materials and methods

4.1. Electrode types

For the experimental part four different types of electrodes were prepared.

(1) Pre-gelled silver/silver chloride electrodes (further referred to as 'gel Ag/AgCl') of type ARBO H92SG from Tyco Healthcare were used as reference.
(2) Dry silver electrodes (dry Ag) with a diameter of 2 cm were cut from a 0.3 mm thin silver foil. They were partly chlorided in an electrochemical process according to Geddes *et al* (1969) to achieve the lowest possible impedance in the actual setup.
(3) To obtain a soft surface as base for the skin adaptive electrode in order to reduce the skin–electrode impedance and motion artifacts, electrodes 2 cm in diameter were punched out of an electrically conductive foam. They were coated with a silver layer 400 nm thick on all surfaces by thermal evaporation to establish an electrical contact similar to that of the dry silver electrodes. A 100 nm layer of titanium was used as an adhesion layer. Different foams were investigated; results for two of them (E103/HART and E103/XAC from STN Schaumstoff Technik Nürnberg) are presented in this paper.
(4) Capacitive electrodes (SiO_2) were fabricated on silicon with a thermally grown silicon dioxide as the dielectric layer. Silicon wafers with doping concentrations varying between 10^{14} atoms cm^{-3} and 10^{20} atoms cm^{-3}, a thickness of the silicon dioxide between 15 nm and 100 nm, and electrode sizes of 1–4 cm^2 were fabricated. The sharp edges of the rectangular chips were protected with adhesive tape so as not to hurt the skin and thus interfere with the measurements.

4.2. Characterization methods

The electrodes were experimentally characterized with respect to the impedance of the electrode–skin interface, the impact of motion artifacts and the function of the integrated passive filter network. These investigations are based on an engineering, not a clinical approach.

4.2.1. Impedance spectroscopy. The electrode–skin contact impedance was analyzed by impedance spectroscopy. The gel electrodes were attached to the skin of the left forearm of the test person using their self-adhesive properties. The test electrodes were attached with a Velcro strap and exchanged carefully between each measurement to avoid any change of the skin surface. The attachment mechanism was adjusted from Searle and Kirkup (2000) by measuring one pair of test electrodes at the time. The pressure of Velcro straps was kept constant for all experiments to ensure the comparability of the results. The skin of the test person was once cleaned by gently wiping it with a 2-propanol impregnated cotton pad, which was allowed to evaporate before applying the electrodes.

The electrodes were connected to a computer-controlled HP4192A impedance analyzer. In order to guarantee reliable and reproducible results, the signal voltage was set to 1 V and the frequency range from 30 Hz to 100 kHz. Even though the recorded impedances will not, due to the restrictions in the test setup in our facilities, equal the absolute values relevant in the ECG measurements, their relative performance can safely be extrapolated to the lower frequency and voltage range in order to predict the relative performance of the different electrodes investigated in this paper. This becomes obvious from the graphs in section 5.

4.2.2. Motion artifacts. The motion artifacts were evaluated from ECGs taken with a long-term ECG recorder, the CardioLight Smart Reader from Medset, Hamburg. This system is designed for gel electrodes and allows recording the ECG for up to 2 days. The different electrodes were attached and exchanged as already described above. The electrodes were positioned under the nipples of the test person close to each other. A reference gel electrode was used, as required by the CardioLight Reader. The measurement can, however, also be performed with a two-electrode setup alone. Motions of the upper body caused only slight zero-line fluctuation in the signal. To evaluate the artifact generation, walking motions were performed to stimulate the electrodes in a natural way. The intensity of the walking motion was increased until the signal of the test electrodes was affected. The measurement started 5 min after the application of each test electrode and lasted for about 5 min.

4.2.3. Passive filtering. In order to validate the theory of the passive filter network, its transfer function was measured in a two-port measurement setup, and results were confirmed with an ECG recording.

The input signal was introduced via two gel electrodes applied to the left forearm. Gel electrodes were used, as they exhibit the best transmission characteristic, i.e. the applied signal passes the interfaces and tissue with low attenuation and their contribution to the transmission signal is minimal. Between these, the electrodes to test were placed and fixed with a Velcro strap. The distance between the gel electrodes was 16 cm, and 3 cm between the gel and the test electrodes. A sinusoidal signal of 2 V peak to peak was applied, and the frequency was swept between 0.1 Hz and 1 kHz using an HP3325A synthesizer/function generator. The output signal was measured with a Tektronix TDS 3012B oscilloscope.

The suppression of the slowly moving offset potentials was confirmed by an ECG recording. An electronic circuit was developed, which, in contrast to standard ECG amplifiers, can cope with the very different input impedances of the two electrodes used for this experiment (Ag and SiO_2). In figure 4, the electrodes are directly connected to an instrumentation amplifier. The ECG signal is amplified by the factor of 100 and fed into a low-pass filter to suppress frequencies beyond 300 Hz. This type of filtering was not sufficient to suppress noise. To eliminate the 50 Hz noise, a shielded measurement setup and symmetric input impedance at an amplifier with high common mode rejection are inevitable. The reference potential of the

Figure 4. Simplified block diagram of an ECG recorder. The electrode leads are directly fed to the input amplifier, and filtered to obtain the ECG signal.

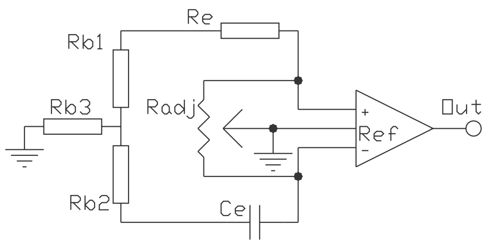

Figure 5. Input circuit of the operational amplifier used in the filter electrode setup using only two electrodes. The adjustable resistor R_{adj} balances the potential instead of using a third electrode as a reference.

instrumentation amplifier has to be adjusted to achieve this symmetry. In a standard recording a third electrode picks up the reference potential. As the arrangement for the filter electrodes uses only two electrodes, an adjustable resistor R_{adj} in the input circuit (figure 5) balances the potentials to the body potential to minimize the 50 Hz noise. The adjustment of the resistor may be annoying, and electronics for zero-point calibration should be implemented in the next version for symmetrizing the impedance. The original motivation behind the design for a two-electrode setup is the miniaturization of the complete ECG-monitoring equipment for added patient comfort (Karilainen *et al* 2004). The amplifier common mode reflection ratio was not optimized in this work. However, its performance was good enough to verify the filter function of the electrode setup.

Again the electrodes were applied with self-adhesive and Velcro strap, respectively, right under the nipples of the test person, and the settling time of the electrodes on the skin as well as the duration of the measurement were 5 min each.

4.2.4. Minimum distance for electrodes. In order to reduce the overall size of the ECG measurement setup in Holter monitoring systems, the electrodes should be as small as possible and positioned on the patient's skin as close as possible (Karilainen *et al* 2004). To investigate the critical distance at which an ECG measurement would still be possible, two electrodes were placed next to each other as close as 1 cm right under the left nipple, where the potential gradient of the *R*-spike is at its maximum. The gel electrodes and the filter electrodes were tested. The gel electrodes were applied to the skin with adhesive and the filter electrodes with a Velcro strap, as described in previous sections.

5. Results and discussion

5.1. Impedance measurements

The skin conductivity varies all over the body (Mayer-Gindner *et al* 2004) especially due to a variation of the properties of stratum corneum or of sweat glands and ducts. Variation is more

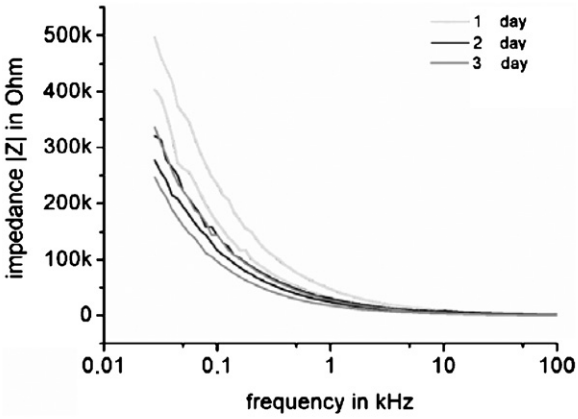

Figure 6. Reproducibility of impedance measurements on various measurements on the same day and on consecutive days. The variation between the measurements on 1 day is smaller than the variation on following days.

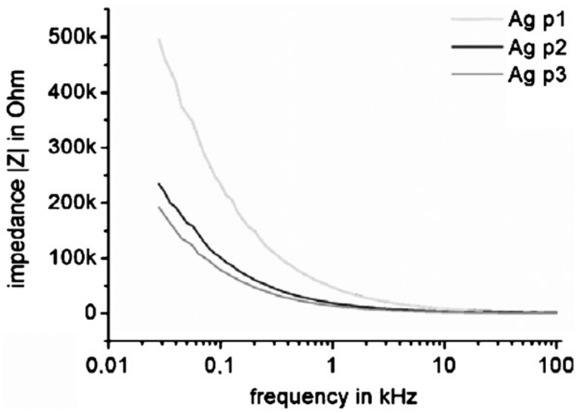

Figure 7. Pressure dependence of skin–electrode impedance. The impedance is strongly dependent on the applied pressure.

prominent for dry electrodes than wet gel electrodes (Yamamoto *et al* 1988). Skin humidity changes from day to day and during the day. Showering or skin care products will moisturize the skin; sun or dry air in the winter will dehydrate it. The resulting changes in skin impedance affect the measurement decisively. Therefore the reproducibility of the measurements was investigated, as well as the influence of pressure on the electrode and its position. Figure 6 demonstrates the impact of these parameters for an Ag electrode applied at the same location on three different days. Differences on 1 day usually are less than for different days. To keep the skin properties as constant as possible, all further measurements were performed during 1 day.

The slight effect of pressure of the electrode on the impedance is shown in figure 7. The electrodes were attached to the forearm at just enough pressure (p1) to ensure a good contact, then at medium pressure (p2) and finally at strong pressure (p3). To achieve comparable results all the following measurements were made at pressure p1.

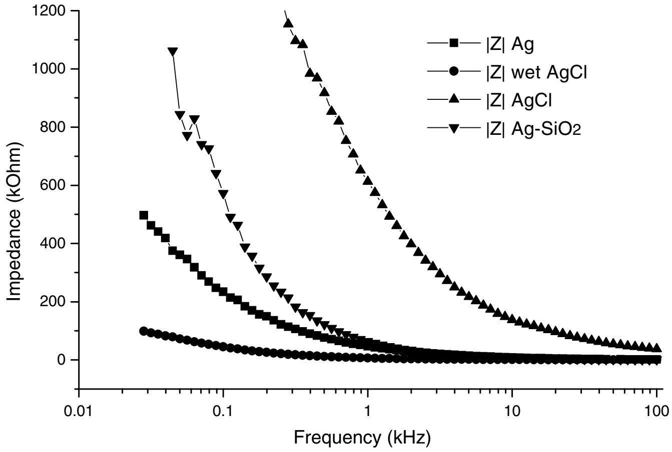

Figure 8. Impedance of tested stiff electrodes. The gel electrode has the lowest impedance because of the moist stratum corneum. The AgCl electrode in the dry condition isolates, producing the large impedance also in the high-frequency region.

The influence of the electrode position on the impedance was also investigated. Since the different distances between the electrodes showed negligible variation due to the high conductivity of the underlying tissue, the results are not depicted.

5.1.1. Impedance properties of rigid electrodes. The impedance of dry Ag/AgCl electrodes, dry Ag electrodes and the combination of an Ag electrode with a SiO_2-on-Si electrode (in the following Ag–SiO_2 electrode) was investigated and compared to that of the Ag/AgCl gel electrodes. As seen in figure 8, the gel electrode exhibits the lowest impedance, because the gel moisturizes the stratum corneum, which increases its ion conductivity. The dry Ag electrodes exhibit higher contact impedance since the stratum corneum keeps its original properties. Because of the lack of moisture, the dry Ag/AgCl electrode performs worst. The chloride isolates in dry condition, which is also evident in figure 8; the high-frequency impedance is the largest of all electrodes. Introducing a 0.9% sodium chloride solution proved that on a liquid interface the Ag/AgCl electrodes exhibit even lower impedance than the commercial gel electrodes. This demonstrates that the skin impedance dominates the skin–electrode impedance. The Ag–SiO_2 electrode exhibits a higher contact impedance than the Ag electrodes due to the SiO_2 capacitor.

5.1.2. Impedance of foam electrodes. To characterize the impedance of the foam electrodes, they were applied to the left forearm, first on the inner side without hair, then on the outer, hairy side of the arm. The results are compared to rigid Ag electrodes at the same positions as reference to investigate to what extent the contact is improved.

The behavior of two foams is discussed in the paper. Their properties are summarized in table 1. Other foams proved for various reasons to be unsuitable for this application.

E103/HART is a relatively rigid foam with open pores, as can be seen in figure 9(a). Its contact impedance on hairy skin is lower at low frequencies than at high frequencies (figure 10). On hairless skin nearly constant contact impedance is found. Presumably the rigid, even sharp edges of the foam may slightly scratch the skin and thus reduce the skin impedance.

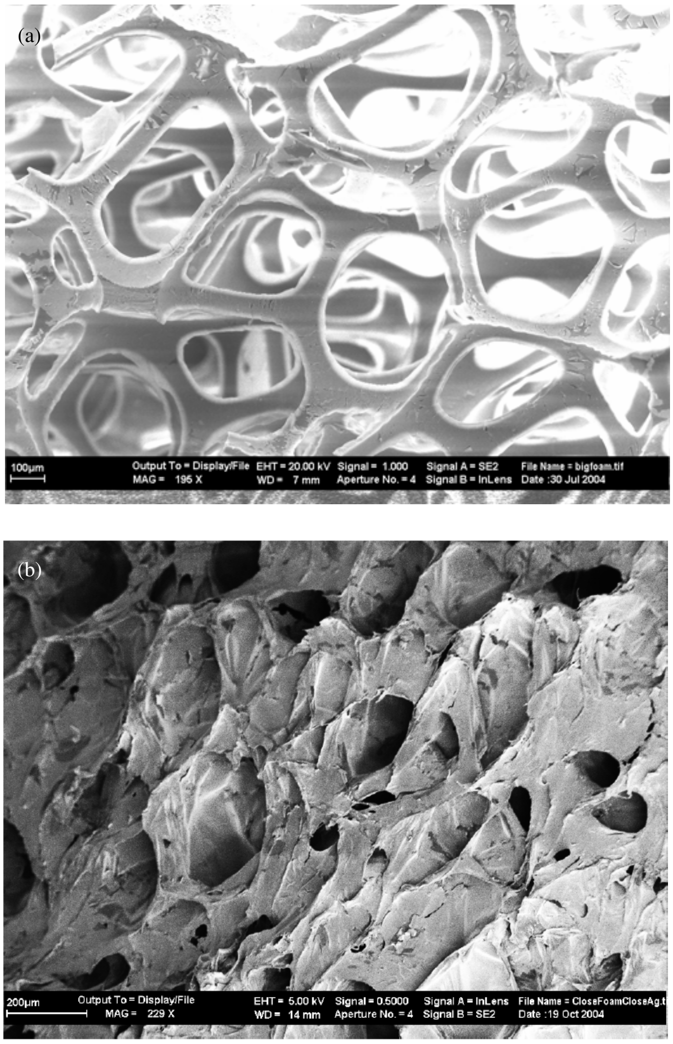

Figure 9. SEM image of (a) E103/HART foam with open pores and (b) E103/XAC foam with closed pores.

Table 1. Properties of tested foam materials.

Foam	Resistivity (Ω cm)	Thickness (mm)	Pores	Material	Bulking 40% (kPa)	Suitability for dry low impedance electrodes
A110/XAC-40	10^7–10^8	5	Closed	Polyethylene	90	Too rigid, not skin adaptive
E103/23	10^3–10^4	3	Open	Polyether	2.5	Too soft, not skin adaptive
E103/HART	10^3–10^4	5	Open	Polyester	60	Suitable
E103/XAC	10^3–10^4	3	Closed	Polyethylene	60	Suitable
E103/XAC-50	10^6	5	Closed	NA	60	Too rigid, slips over skin

Figure 11 shows a further reduced impedance of the E103/XAC foam with closed pores (figure 9(b)). On the hairy side of the arm the contact impedance nearly equals the impedance

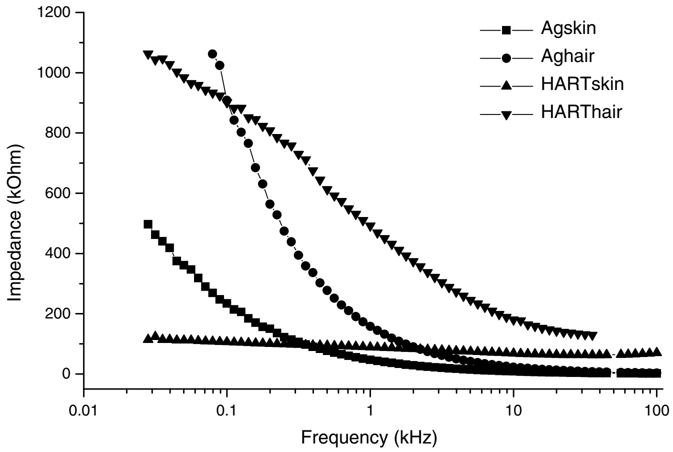

Figure 10. Frequency characteristic of E103/HART foam electrodes. The impedance is possibly influenced by the hard foam slightly scratching the skin.

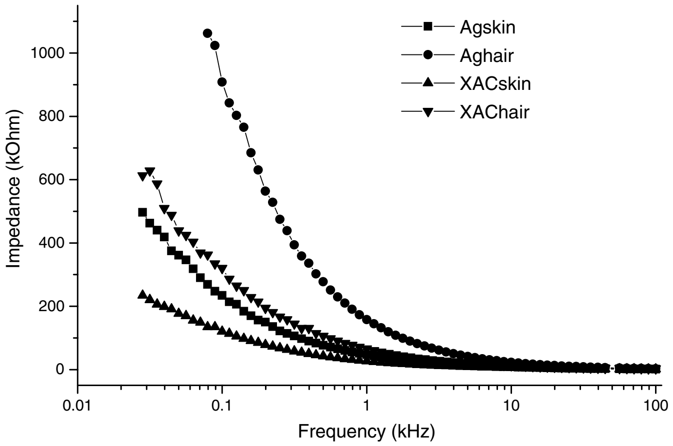

Figure 11. Frequency characteristic of E103/XAC foam electrodes. The foam electrode exhibits a lower impedance than the metal-plate electrode both on hairy and on hairless skin.

of the silver-plate electrode on hairless skin. On hairless skin the impedance is even lower. Evidently the foam is soft enough to contact the skin properly, and the silver layer is very stable. These properties make the standard preparation, i.e. shaving the patient's chest, unnecessary. Certainly, dry electrodes will hardly surpass the properties of gel electrodes. Their performance, however, is comparable in the performed experiments, and a superior long-time stability can be expected due to the fact that no gel, apt to drying, is necessary. In summary, for E103/XAC foam electrodes an impedance comparable to that of Ag/AgCl gel electrodes is achieved.

5.2. Motion artifacts

With gel and Ag-plate electrodes applied, the intensity of walking motions was increased until one of the signals became distorted. The gel electrodes showed the motion artifacts first

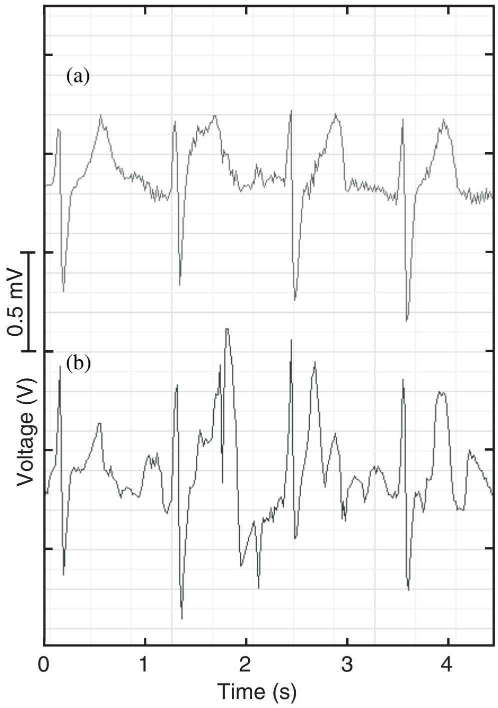

Figure 12. Comparison of motion artifact sensitivity of (a) dry silver electrode and (b) wet gel electrode. Dry electrodes are more resistant to motion artifacts.

(figure 12), i.e. the motion artifacts can be reduced by using dry electrodes, as already discussed in Searle and Kirkup (2000).

The E103/HART and E103/XAC foam electrodes exhibit a slightly better behavior than the gel Ag/AgCl electrodes. A direct comparison of the foam electrodes to the silver-plate electrodes (figure 13) shows that the foam electrode can reduce motion artifacts further, due to its mechanical softness and a better adhesion to the skin.

From the different foams investigated rigid electrodes from E103/XAC and E103/HART adapt optimally to the skin and produce the most intimate contact. Under strain only the inner parts of the electrode move, so there is no friction at the interface. Optimizing the application mechanism will further improve their properties. Foam embedded into a rigid cup with a self-adherent surface similar to a commercial gel electrode would enable the application at constant pressure and avoid the friction against the skin completely. For all foam electrodes the adhesion of the silver layer will have to be improved. Measurements without body motions showed a smooth ECG signal for all electrodes, except for the 50 Hz noise, due to the ECG recorder not being optimized for dry electrodes.

5.3. Capacitive electrodes

Figure 14 shows the attenuation of the gel and the dry electrode compared to the Ag–SiO_2 filter electrode setup. A strong attenuation is observed at low frequencies, i.e. the setup works as a high-pass filter. The high impedances of the SiO_2 and Ag electrodes dominate the transmission function, as compared to the gel electrodes, which apply the signal.

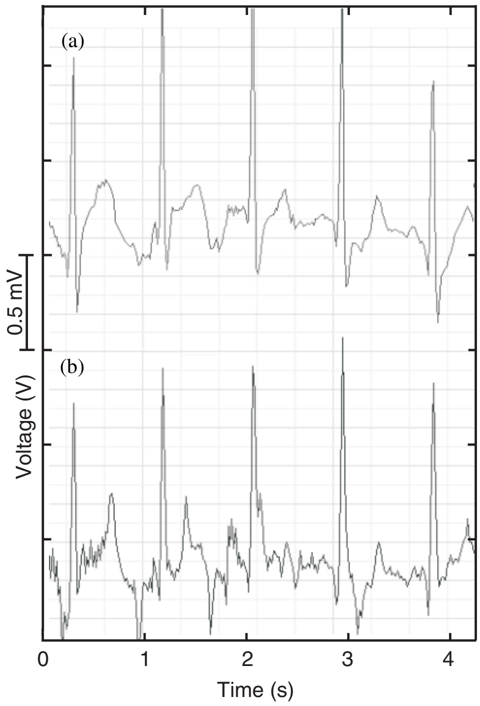

Figure 13. Comparison of motion artifact sensitivity of the (a) soft foam electrode and (b) dry silver electrode. The soft electrodes can reduce the motion artifacts even more than dry electrodes.

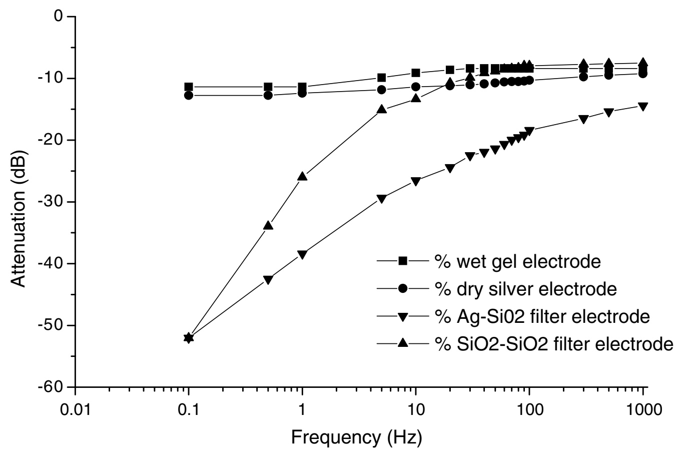

Figure 14. Attenuation of the filter electrodes as compared to dry and wet ohmic electrodes. Strong attenuation can be observed at low frequencies with the filter electrode setup.

A variation of the SiO_2 electrode area between 1 and 4 cm^2 showed no significant impact on the transmission function. The same holds for a thickness variation of the isolator because the oxide layer is much thinner than the isolating layer of the skin. A variation of the conductivity of the silicon from lightly doped (10^{14}–10^{15} cm^{-3}) to highly doped silicon (10^{17}–10^{20} cm^{-3}) showed no significant influence either. The high resistance of the skin is again dominant. These results are not depicted in order to not add pages in the paper.

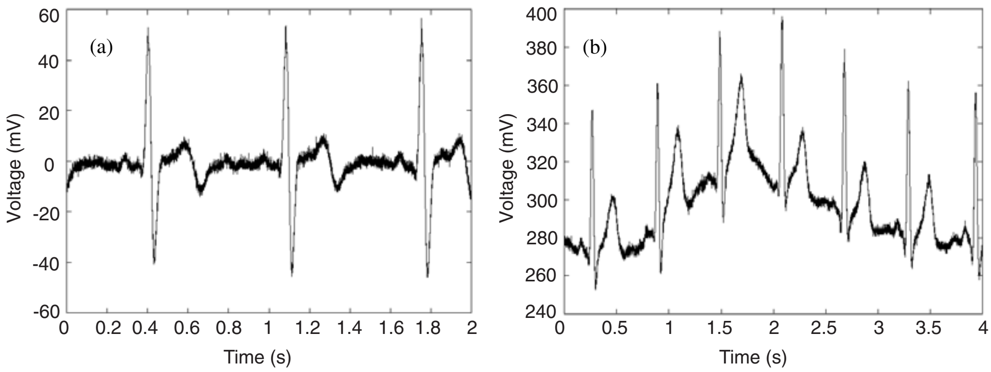

Figure 15. (a) AC-mode and (b) dc-mode measurements of wet gel electrodes. Zero-line fluctuation caused by breathing can be seen in the dc measurement.

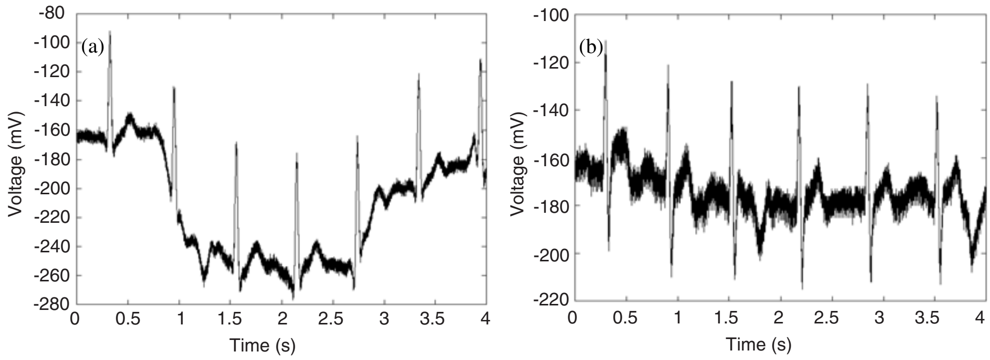

Figure 16. DC-mode measurements with (a) dry Ag electrodes and (b) Ag–SiO_2 electrode setup. The zero-line fluctuation is suppressed by the filter electrodes.

When two SiO_2 electrodes are used, the filter behavior improves. Figure 14 compares this improved characteristic to the Ag–SiO_2 electrode setup. From the measured transmission function the cutoff frequency of less than 5 Hz is derived. A cutoff frequency of 5 Hz is probably not low enough for recording an ECG with a very high resolution, as it is within the band frequency of interest, but can be decreased by some changes in the electrode design.

Figure 15(a) shows the ECG signal measured in the ac mode with gel electrodes. A zero-line fluctuation is not visible, as it is suppressed in the ac mode of the oscilloscope. The cardiac signal with cycle duration of 800 ms is clearly visible. The QRS complex has the peak-to-peak amplitude of about 80 mV. Figure 15(b) shows the zero-line fluctuation caused by breathing. The offset voltage is caused by the filter network and the offset of the input signal, which can be evaluated at the output of the instrumentation amplifier.

Apart from the inferior noise suppression, the results for dry Ag electrodes (figure 16(a)) about equal those of the gel electrodes. The increased noise is due to the local variation of the skin properties. The dc measurement shows a zero-line fluctuation comparable to that of the gel electrodes.

The measurement with the Ag–SiO_2 electrodes (figure 16(b)) is not straightforward because of the difference in the impedances, which introduces excessive noise. Adjusting

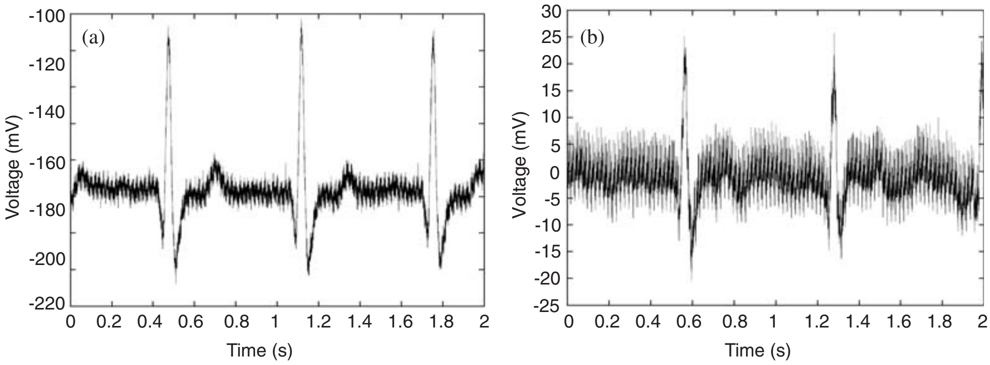

Figure 17. (a) Gel electrodes and (b) Ag–SiO_2 electrodes placed next to each other at a selected position. Positioning the electrodes as close as 1 cm from each other still produces a usable signal.

R_{adj} of figure 5 and adding a 50 Hz notch filter makes the signal detectable. The remaining noise is due to harmonics of 50 Hz, which can be suppressed by adding further notch filters. With an amplitude of about 80 mV the coupling is similar to that of the ohmic electrodes. The dc measurement, however, demonstrates the pronounced zero-line suppression as predicted from the impedance measurements, i.e. overpotentials are completely removed from the ECG signal. The advantage of this electrode setup is that the spurious signal is filtered directly at its origin, and the overloading of the input amplifier will be avoided.

5.4. Minimum distance for electrodes

The gel electrodes and the filter electrodes were positioned as close as 1 cm on the chest, where the heart signal is strongest. The gel electrodes and the filter electrodes were tested. The gel electrodes were applied to the skin with adhesive and the filter electrodes with a Velcro strap. The gel electrodes still generate a usable ECG signal (figure 17(a)). Ag–SiO_2 electrodes show an equivalent response with the expected increased noise (figure 17(b)). Higher amplification and improved passive and active filtering, however, can solve this problem. By finding the trade-off between the separation of the electrodes and the size and energy consumption of the Holter recorder, a miniaturized measurement device can be realized with increased patient comfort.

6. Summary and outlook

New dry and capacitive electrodes for monitoring of the heart function were investigated. The new electrodes avoid the shortcomings of standard Ag/AgCl gel electrodes. The standard gel electrodes are susceptible to dehydration, thus having unstable impedance, and they support bacterial growth. Three novel concepts for dry electrodes to improve such standard electrodes were investigated. Rigid silver plates, silver plates coated with silver chloride, Ag-coated conductive polymer foam soft electrodes, and capacitive SiO_2–Si electrodes were designed, fabricated and characterized with the objective of improving the contact on hairy skin to reduce the electrode impedance, to diminish motion artifacts and to passively filter zero-line fluctuations.

By impedance spectroscopy of the electrode–skin interface it is demonstrated that dry foam electrodes exhibit impedance properties comparable to those of the standard gel electrodes both on hairy and hairless skin. Thus shaving and abrading, as recommended for standard gel electrodes, can be avoided. Also motion artifacts in the ECG measurement can be suppressed effectively by choosing the appropriate foam.

With a combination of an Ag and a capacitive SiO_2–Si electrode a passive filter network to suppress zero-line fluctuations was demonstrated. With some adjustments in current ECG recorders to deal with the different electrode impedances, this arrangement could be adapted for use.

Dry capacitive electrodes are suitable for short-term ECG measurement, as also discussed in Prutchi and Sagi-Dolev (1993), but such electrodes are susceptible to artifacts, as they are hard and can slip over the skin, which causes loss of contact and charging effects. Also it is questionable whether wearing such electrodes is possible during longer recordings. Silicon dioxide used in this work is very thin, but has an extremely high capacitance, which ensures high coupling of the signal. It has been argued that corrosion of the dielectric of capacitive electrodes would pose a risk for their use, but constructing a foam cushioning layer, as discussed above, on top of the SiO_2 electrode to combine both the electrode–skin impedance reduction and artifact reduction, as well as the passive filtering of offset potential properties, will result in an improved ECG monitoring without the risk of wearing of the extremely thin oxide layer.

Next, it must be tested whether the characteristics of electrodes described in this paper are stable in long-term use. It will be investigated whether other problems, such as bacterial growth, arise when the electrodes are in contact with the skin for several days or even weeks. Furthermore, electronics for zero-point calibration for the filter electrode measurement device will be implemented in the next version for symmetrizing the impedance.

Future work will concentrate on the development of a soft capacitive electrode to combine the advantages of both new types of electrodes for a long-term ECG system, which is convenient with respect to all relevant electrode properties. The miniaturization and simplification of the electrode attachment, by combining two electrodes next to each other on one system, will be pursued. The processes to deposit Ag films on foams with a strong abrasion resistance have to be optimized. A mechanical stabilization of the foam in a rigid cup-like and self-adhesive package for a more defined, long-term stable contact to the skin must be developed.

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