Dry and Capacitive Electrodes for Long-Term ECG-Monitoring

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Abstract— **In this paper the development, fabrication and characterization of new dry and capacitive bioelectrodes for long-term biopotential monitoring is presented. The electrodes improve the applicability of dry electrodes in long-term recording of ECG by reducing the contact impedance between the electrodes and the skin as well as artifacts induced due to motions of the body. 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 such as breathing or changes in the electrode-skin interface properties.**

The new dry and capacitive electrodes have an impedance and generate signals comparable and superior to those of presently used standard gel-electrodes at significantly improved long-term stability. The integrated filter network effectively eliminates fluctuating offset potentials.

Index Terms— **Artifact, Capacitive Electrode, Dry Electrode, ECG, Impedance**

I. Introduction

Rapid advancement in medical technologies in the past have contributed to significant improvements in patient care. Partly because of the 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 of health care have increased due to novel more expensive medical treatment. The challenges for engineers are to develop new or improve the methods of preventive care and decrease the costs of instrumentation as well as of personnel and maintenance. Especially microsystem technologies offer numerous ways to generate miniaturized medical systems, since material costs and reliability can be superior to other technologies. Furthermore, miniaturizing such systems increases the patient comfort considerably [1].

Cardiovascular diseases are the main cause of death within the population in the age of 44 - 64 years, and the second most frequent cause of death of people between 24 and 44 years. In Germany e.g. about 300 000 people suffer from a heart attack annually. An early recognition of attack symptoms and warning of the patient or doctor would enable preventive actions to avoid the attack and thus reduce the risk of irreparable damage to organs, or even death. Monitoring risk groups, such as people who recently were subject to a bypass surgery or pacemaker implantation, has proven to effectively decrease the number of heart attacks. Long term recording of ECG (electrocardiogram) is a standard procedure in current cardiac medicine, but the devices are capable of monitoring the heart function for a time period of only a few days, whereas much longer recording times are of clinical interest. Longer recording times are mainly limited by the restrictions in the electrode performance.

The objective of this work was to generate a new type of a dry, long time applicable ECG electrode, which achieves a reduction in contact impedance and motion artifacts as compared to conventional dry electrodes. Different types of dry conductive and capacitive electrodes for the use in long term monitoring of heart function and other biopotentials were designed, fabricated, and characterized. Furthermore, a method for a passive suppression of slowly changing offset potentials was developed, which uses a special electrode configuration.

II. Biomedical Basics

Biopotentials are electrical potentials inside a living body which are created by ionic currents. 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 called 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 instable, as it depends on skin properties such as humidity and thickness. Typical values can be as low as 100 nF per cm^2 [2]. 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 half-cell potentials caused by different energies of the electrode, electrolyte and skin. All of these contributions can be described by an electrical equivalent circuit depicted in Fig. 1,

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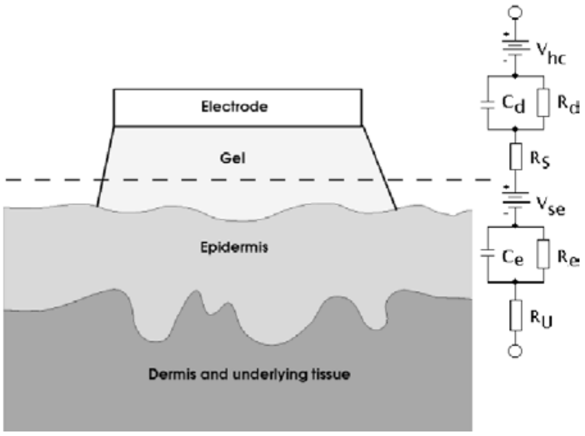

Fig. 1. Electrical equivalent circuit of skin-electrode system.

as proposed in [3]. The task of the electrolyte gel of gel electrodes is to reduce the influence of the skin on the impedance by moisturizing its dry outer layer and make it highly ion-conductive. When dry electrodes are used, moisturizing is significantly reduced, even though perspiration under the electrode has a similar effect.

Depending on the location on the body, different ECG signals of different amplitudes are obtained. Since the measured signal is a superposition of all electrical activities in the body-electrode setup, extra cardiac muscle movements such as walking will cause motion artifacts which disturb the signal. A very evident disturbance to the ECG signal are electrodes which loose their contact. This event pulls the signal beyond the measuring range. Another artifact observed in ECG measurements is the zero line fluctuation caused by slow motions such as breathing. When the lungs fill up with air, the position of the heart changes relative to the electrodes, which in turn changes the voltage between them. Furthermore, stretching the skin under the electrode changes the electrical properties of the skin and the interface, which may cause further artifacts in the signal. During the analysis of an ECG signal all these phenomena have to be accommodated to avoid misinterpretations.

The standard efficient transducer is the non polarizable Ag/AgCl gel-electrode, which exhibits only the ohmic resistances R_d and R_s in the equivalent circuit. Dry electrodes on the other hand exhibit both polarization and conductivity, they are thus partly polarizable. Thus they can be modeled by a parallel circuit of a dry ohmic and an insulated capacitive electrode. The capacitive electrode is modeled as perfectly polarizable, i.e. its equivalent circuit is just a capacitor C_d.

The standard Ag/AgCl gel-electrodes have limited shelf life and are not reusable. They can be used only for a few days because they suffer from dehydration. Dehydration leads to a modified electrode impedance, which generates noise and other artifacts in the measured signal. Furthermore, the gel can cause skin irritations and support bacterial growth [4].

It has been proposed, that the long term performance of an ECG setup could be improved by using dry and capacitive electrodes, instead. Even though also positive results have

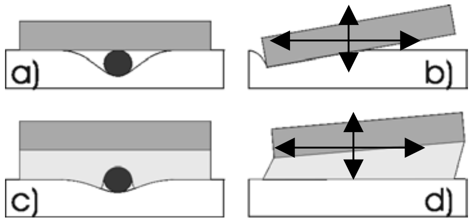

Fig. 2. Advantages of foam electrodes in comparison to stiff dry electrodes. a) Loss of contact area and increase of contact impedance due to hairs. b) Loss of contact and increase of artifacts due to charging effects caused by motion of electrode relative to skin. c) Increased the contact area. d) Contact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

been published [4, 5], known dry electrodes often generate evident problems such as lack of contact on hairy skin, higher contact impedances between the body and the electrode and charging of the electrodes by friction against the skin. Attempts to overcome these problems include invasive techniques where the skin is mechanically punched to lower its impedance [6]. Even this mechanism is not long-term stable since the skin regenerates and encapsulates the electrode in the saline layers of skin by natural wound healing. Furthermore, invasive techniques possess a high risk of inflammation.

III. Novel Dry and Capacitive Electrodes

Present dry electrodes are metal plates, the contact impedance of which to the skin is significantly higher and less defined than that of gel electrodes because of reduced moisture and undefined contact area due to its rigidness and locally isolating hairs between electrode and skin. In ambulant monitoring shaving the patient before applying the electrodes is arduous, and in long term monitoring even shaving does not produce stable results because the hair regrows within a few days. Also the standard preparation method in ambulant care, a slightly invasive mechanical abrasion of the skin, does not solve the problem because the skin regenerates within about 24 hours. A shift of the rigid electrodes relative to the skin caused by unavoidable patients` motions during long time monitoring results in random variations of electrode contact area as well as charging which especially impair capacitive electrodes.

To overcome all these problems a novel soft dry electrode was realized, which adapts to the skin topography such as curvatures and hairs, and experiences low relative motion of skin to electrode because of its flexibility and cushioning effect (Fig. 2). In addition, a novel concept to suppress unwanted slow sub-Hertz offset potential fluctuations due to potential changes caused by a drift in the electrode properties or patient motions like breathing was developed: a passive filter network integrated into a novel electrode setup acts as a passive low-pass filter directly at the origin of the signal.

A. Reduction of Contact Impedance

The electrode-skin interface impedance is governed by contact area and skin properties. In order to obtain a sufficiently low impedance the contact area should be large.

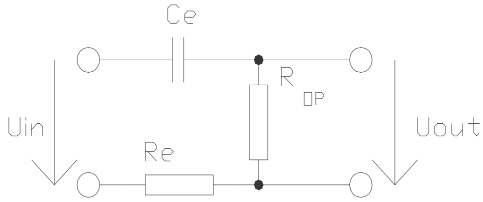

Fig. 3. Schematic presentation of passive filter electrode network comprising of ohmic and capacitive electrodes, as well as the input impedance of the operation amplifier.

On hairy skin, air gaps hundreds of micrometers wide can occur between the electrode and the skin, which will significantly increase this impedance. A skin adaptive electrode e.g. a conductive foam, soft enough to adapt the geometry of the hair, would reduce this effect (Fig. 2 c). As long as the hair does not completely conceal the skin, foam-electrodes can still contact the skin, whereas rigid electrodes will fail even at loose occlusion. Under slight pressure, the contact of foam-electrodes on hairy skin is nearly as safe as that of rigid electrodes on hairless skin. Skin conductivity and the permeability can only be changed by moisturization or destruction of the stratum corneum. In a real application, however, during long term measurements the skin under the electrode will be moisturized by sweat and thus the interface impedance will reduce. In spite of the lower ion conductivity of sweat as compared to a gel the interface impedance becomes low enough for a safe signal detection.

B. *Reduction of Motion Artifacts*

The artifacts generated in biopotential electrodes due to motion are induced by charging because of friction and slipping, as well as by disturbance of the electrical double layer at the interfaces. Since materials 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 restoration to equilibrium is nearly instant in metal, but slow in a dielectric which in turn will induce charges in the metal in a separation-contact-sequence. A soft electrode made of foam will not be subject to this effect, since the contact even under motion is maintained. Also rubbing and sliding of the electrode on the skin will hardly induce any charge. The attachment mechanism creates a pressure to the electrode and its elasticity improves the contact both horizontally and vertically (Fig. 2 d). Furthermore, a deformable electrode reduces its motion relative to the skin. Even if motion occurs, only parts of the electrode move, as the walls of the pores are not rigidly connected to each other. Thus only parts of the electrode will rub on the skin i.e. the amount of generated charges is reduced.

C. *Dry Electrodes as Passive Filter Network*

One cycle of the heart function takes about one second. The QRS-complex, which lasts less than 0.1 s, contains the highest frequency. Thus, for ECG, frequencies from 1 Hz up to 250 Hz are of clinical relevance. Spurious signals such as the zero line fluctuation, which are in a frequency range below 1 Hz are suppressed by a high pass filter, which is commonly integrated into the signal amplifier. Since the height of these fluctuations often surpass the heart signal significantly, it would be advantageous to suppress them at their origin, i.e. to supply the electrodes with a passive filter network. Combining ohmic and capacitive electrodes (Fig. 3), such a filter network can be integrated into the transducer.

IV. Materials and Methods

A. *Electrode Types*

For the experimental part four different types of electrodes were used. Pre-gelled silver/silver chloride electrodes of type ARBO H92SG from Tyco Healthcare were used as reference. Dry silver electrodes 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 the parameters suggested in [7] to achieve the lowest possible impedance in the actual setup. To obtain a soft surface as base for a skin adaptive electrode in order to reduce the electrode-skin contact impedance and motion artifacts, an electrically conductive foam was used. The electrodes were fabricated by punching circular pieces 2 cm in diameter out of the foam. The punch outs were coated with 400 nm thick silver layer in an evaporation process. A 100nm 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 discussed in this paper. Capacitive electrodes were fabricated on silicon with a thermally grown silicon dioxide as the dielectric layer. Electrodes with various doping concentrations, thickness of the dielectric, and electrode sizes were fabricated.

B. *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.

1) *Impedance Spectroscopy*

The electrode-skin contact impedance of the electrodes was analyzed by impedance spectroscopy. To characterize the properties of a frequency dependent impedance, current is measured while applying a sinusoidal voltage in the frequency range of interest. During all measurements, the gel-electrodes used a self adhesive attachment mechanism, whereas the other electrodes were attached with a Velcro strap to the skin of the left forearm, cleaned by gently wiping it with a 2-propanol impregnated cotton pad. The electrodes were connected to a computer controlled HP4192A impedance analyzer. Signal voltage for all measurements was 1 V. The frequency ranged between 30 Hz and 100 kHz.

2) *Motion Artifacts*

The motion artifacts were evaluated from ECGs taken with a long term ECG recorder, model CardioLight Smart Reader from company Medset. This system is designed for gel-

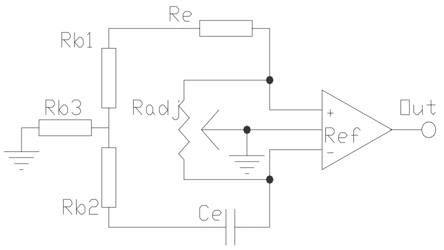

Fig. 4. Input circuit of the operation amplifier used in filter electrode setup.

electrodes and allows to record the ECG for up to two days. The gel-electrodes use a self adhesive attachment mechanism. Other electrodes were attached to the chest with a belt. To evaluate the artifact reduction, 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 foam electrodes, the dry Ag/AgCl as well as the dry Ag test electrodes were compared with the gel-electrodes as the reference. The susceptibility of the capacitive electrodes to motion artifacts was not investigated in this experiment, since they are expected to have the artifact sensitivity in the same range as the silver plate electrodes.

3) Passive Filtering

In order to validate the theory of the passive filter network, its behavior was characterized by measuring its transfer function in a two-port measurement setup and in an ECG recording.

Two port networks are characterized by applying an input signal and comparing it to the output signal. To introduce the input signal two gel-electrodes were 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. Between these two gel-electrodes the tested electrodes were placed. A sinusoidal signal of 2 volts 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 in a newly designed measurement setup, because the input impedances of the two electrodes (Ag and SiO_2) are too different for a standard ECG amplifier, which would result an unreadable ECG signal. Filtering alone is in this case not enough to suppress the 50 Hz noise, but it has to be eliminated at its origin. A symmetric input impedance at an amplifier is inevitable. The reference potential of the instrumentation amplifier has to be adjusted to achieve this symmetry. In a standard recording this is realized with a third electrode which picks up the reference potential. For the measurement with the filter electrodes, however, only two electrodes are used. The problem is solved with the adjustable balancing resistor R_{adj} in the input circuit (Fig. 4), with which the reference potential was manually adjusted to the ground potential of the body.

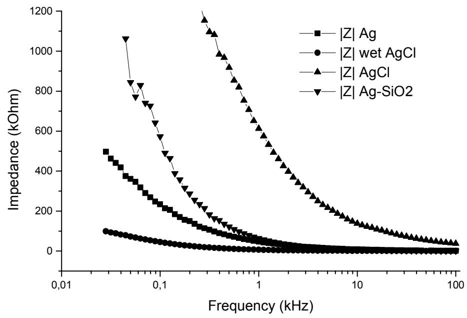

Fig. 5. Frequency behavior of tested stiff electrodes.

V. Results and Discussion

A. Impedance Measurements

The skin conductivity varies all over the body [8] especially due to a variation of the properties of stratum corneum or of sweat glands and ducts. The variation is increased when dry electrodes are used as opposed to wet gel electrodes [9]. 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. Differences in impedance at various times on the same day showed to be smaller than the differences in measurement performed on consecutive days. To keep the skin properties as constant as possible, all further measurements in this work were performed during one day. It was also found out, that the impedance is a function of the pressure with which the electrode is applied to the skin. Pressure will not only improve the contact but leads to a hydration of the lower skin levels and thus to an impedance reduction. In the following measurements the pressure was controlled to eliminate the influence of application pressure on the results. Next, the dependence of the position of the electrodes on the impedance was investigated. Different distances between the electrodes on the left forearm showed negligible variation due to low impedance of the underlying tissue.

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) were investigated and compared to that of the Ag/AgCl gel electrodes. The electrodes were attached to the forearm with a belt as described above. As seen in Fig. 5, the gel-electrode achieves the best properties, because the gel moisturizes the stratum corneum, which increases its ion conductivity and lowers its impedance. The dry Ag-electrodes exhibit a 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 isolating layer of silver chloride between skin and conductive silver creates a conductive interface for ions only if humidity is present. The

TABLE I
PROPERTIES OF TESTED FOAM MATERIALS

Foam	Resistivity (Ω/cm)	Thickness (mm)	Pores	Material	Bulking 40% @ (kPa)	Suitability for dry 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	N.A.	60	Too rigid, slips over skin

Fig. 6. Definition of skin and hair in impedance measurements.

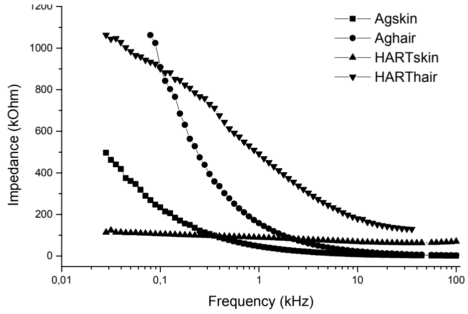

Fig. 7. Frequency characteristic of E103/HART foam electrodes.

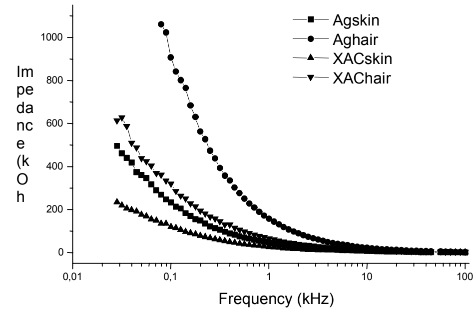

Fig. 8. Frequency characteristic of E103/XAC foam electrodes.

Ag-SiO_2 electrodes exhibit a higher contact impedance than the Ag electrodes due to the SiO_2-capacitor.

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 other, hairy side of the arm (Fig. 6). The results are compared to Ag electrodes at the same positions as reference to investigate to what extend the contact compared to rigid electrodes is improved.

The behavior of two foams are discussed in the paper. Other foams which were investigated for various reasons proved not suitable for this application (see Table 1).

E 103/HART is a relatively rigid foam. Its contact on hairy skin is better in the low frequency range than at high frequencies (Fig. 7). On hairless skin, a nearly constant contact impedance is found. The rigid, even sharp edges of the foam may slightly scratch the skin and thus reduce the contact impedance. Even though invasive methods are not desired, because of additional uncontrollable biological effects which may occur at the electrodes, such a soft mechanical abrasion had no evident negative effect. This abrasion has a less impact than shaving or abrading the skin with a grinding paper, which are both standard preparation methods in current ambulatory practice. The foam E 103/HART reduces the impedance to the same level as the Ag/AgCl gel-electrodes on hairless skin.

Fig. 8 shows a further reduced impedance of the E 103/XAC foam. On the hairy side of the arm the contact impedance nearly equals the impedance of silver plate electrode on hairless skin. On hairless skin the impedance is reduced even further. Evidently the foam is soft enough to contact the skin properly and the silver layer is very stable. This improvement makes the standard preparation, shaving the patient's chest, unnecessary. Certainly, dry-electrodes can not surpass the properties of gel-electrodes. They can, however, reach nearly the same level at an improved long-time stability. In summary, for E 103/XAC foam-electrodes an impedance comparable to that of an Ag/AgCl gel-electrodes is achieved.

B. Motion Artifacts

With gel and Ag-plate electrodes applied, the intensity of motion by walking was increased until one of the signals became distorted. The gel electrodes showed the motion artifacts first (Fig. 9 a, b), i.e. the motion artifacts can be reduced by using dry electrodes, as already discussed in [4].

The E 103/HART and E 103/XAC foam electrodes show a slightly better behavior than the gel Ag/AgCl-electrodes. A direct comparison of the foam-electrodes to the silver plate electrodes (Fig. 9 c, d) shows, that the foam-electrode can reduce motion artifacts further.

The results for the motion artifacts reveal that soft electrodes such as E 103/23 which deform very easily are not optimally skin adaptive, or do not posses the mechanical or chemical properties to support an evaporated metal layer. The very rigid foams such as E 103/XAC-50 are also not skin adaptive, and the metal layers also tend to crack during processing and

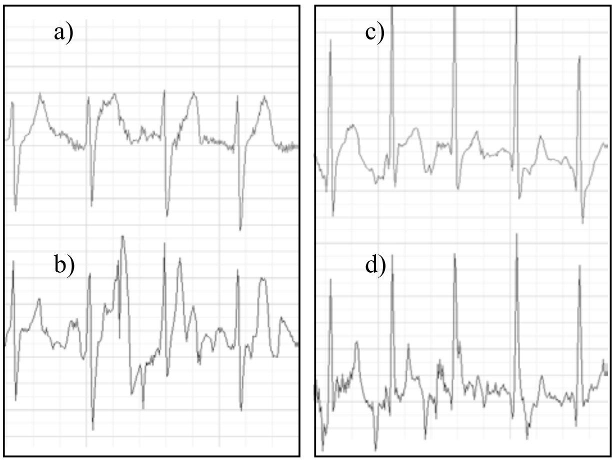

Fig. 9. Comparison of movement artifact sensitivity of a) dry silver electrode and b) wet gel electrode. Comparison of movement artifact sensitivity of c) soft foam electrode and d) dry silver electrode.

use. Optimally rigid electrodes, such as E 103/XAC and E 103/HART, which adapt to the skin under pressure, produce the most intimate contact. Under stress, only the inner parts of the electrode move, and so there is no friction at the interface. Optimizing the application mechanism will further improve their properties. A 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 was somewhat problematic and has 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 optimized for dry electrodes.

C. Capacitive Electrodes

Fig. 10 shows the attenuation of the gel- and of 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. Due to the loose coupling of the SiO_2 and Ag electrodes as compared to the gel electrodes, which apply the signal, they dominate the transmission function.

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. Changing from lightly doped to highly doped silicon has no significant influence either. The high impedance of the skin again is dominant.

When two SiO_2-electrodes are used, the filter behavior improves. In Fig. 10 this improved characteristic is compared to the Ag-SiO_2-electrode setup. From the measured transmission function the cut off frequency is derived. The cutoff frequency in this case is clearly visible at less than 5 Hz.

Fig. 11 a shows the ECG signal measured in the AC-mode with gel-electrodes. A zero line fluctuation is not visible, because in the AC-mode of the oscilloscope the low

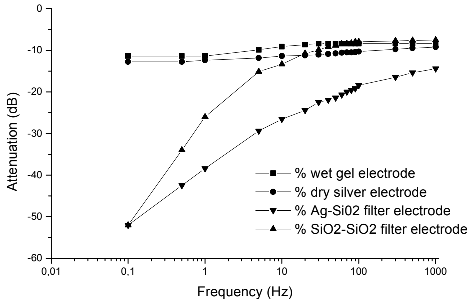

Fig. 10. Attenuation of the filter electrodes as compared to dry and wet ohmic electrodes.

frequencies are suppressed by the input filter. The cardiac signal with a cycle duration of 800 ms is clearly visible. The QRS-complex has a peak to peak amplitude of about 80 mV. Fig. 11 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.

Aside from the higher amplitude (100 mV) of the QRS-complex but more problems with noise suppression, the measurement with dry Ag-electrodes in Fig. 12 a shows nearly the same behavior as the gel-electrodes. The increased noise is due to the local variation of the skin properties. In contrast to dry electrodes gel electrodes have the capability to reduce the impedance by moisturization. The DC-measurement shows a zero line fluctuation comparable to the fluctuation of the gel-electrodes.

The measurement with the Ag-SiO_2-electrodes is not straight forward because of noise which makes the signal nearly undetectable. The different electrodes present two completely different impedances to the amplifier input. Adjusting R_{adj} of Fig. 4 and adding a 50Hz 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 for these frequencies. With an amplitude of about 80 mV the coupling is similar to that of the ohmic electrodes. The DC-measurement (Fig. 12 b), however, demonstrates the effective zero line suppression as predicted from the impedance measurements, i.e. over potentials are completely removed from the ECG-signal.

VI. Summary and Conclusion

We have designed, fabricated and characterized different types of dry and capacitive electrodes for long term monitoring of the heart function. The standard Ag/AgCl gel-electrodes are susceptible to dehydrating, which modifies their impedance, and support bacterial growth. Three novel concepts to improve such standard electrodes were investigated: reduction of the electrode impedance by improving the contact on hairy skin, reducing motion artifacts, and passively filtering the zero line fluctuations. Rigid silver plates, silver plates coated with silver-chloride, Ag-coated conductive polymer foam soft electrodes, and capacitive SiO_2-

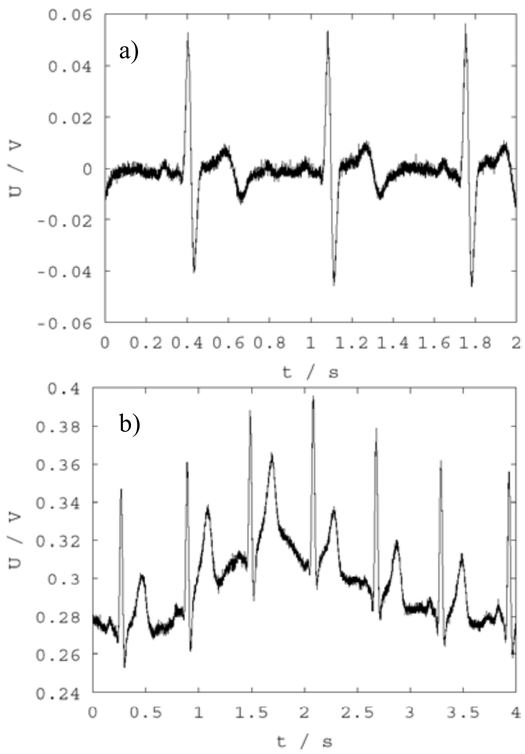

Fig. 11. a) AC and b) DC measurements of wet gel electrodes.

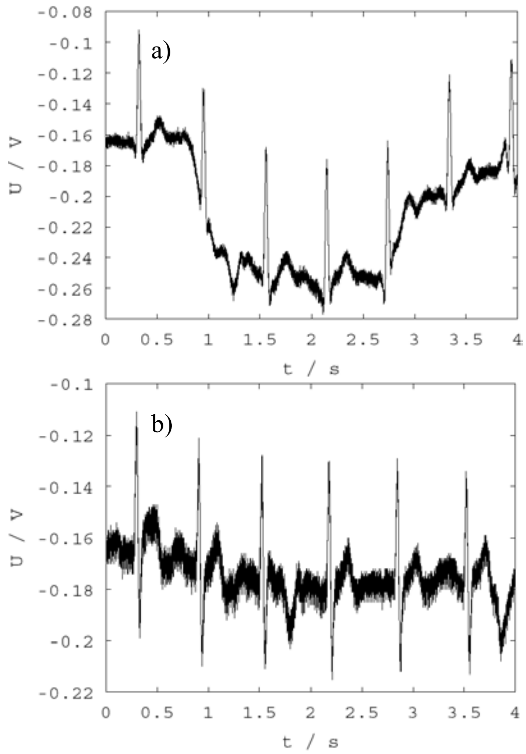

Fig. 12. DC measurements with a) dry Ag electrodes and b) SiO_2-Ag electrode setup.

Si-electrodes were fabricated and characterized.

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 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 is also suitable for long term measurement.

Dry capacitive electrodes are suitable for short term ECG measurement, 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. Silicon dioxide used in this work is very thin, and 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 potentials properties, will result in an improved long term ECG monitoring without the risk of wearing of the extremely thin oxide layer.

Future work will concentrate on the development of such 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 mechanical stabilization of the foam in a rigid self adhesive surface for a more defined long term stable contact to the skin has to be developed.

References

[1] A. Karilainen, T. Finnberg et. Al.. Mobile Patient Monitoring Based on Impedance-Loaded SAW-Sensors. IEEE Transactions on Ultrasonics, Ferroelectrics and Frequency Control. Vol. 51. No. 11. Pp 1464 – 1469. 2004.

[2] Taheri, B. A., Knight, R. T., et. Al. A dry electrode for EEC recording. Electroencephalography and Clinical Neurophysiology. Vol. 90. Pp. 376 – 383. 1994

[3] J. G. Webster, Medical Instrumentation, Application and Design. New York: Wiley. 1998. P. 198.

[4] A. Searle and J. Kirkup. A direct comparison of wet, dry and insulating biolelectric recording electrodes. Physiological Measurement. Vol. 21. Pp. 271 – 283. 2000.

[5] D. Prutchi and A. Sagi-Dolev. New Technologies for In-Flight Pasteless Bioelectrodes. Aviation. Space and Environmental Medicine. Vol. 64. Pp. 552 – 556. 1993.

[6] P. Griss, P. Enoksson et. Al. Micromachined Electrodes for Biopotential Measurements. Journal of Microelectromechanical Systems. Vol.10. Pp. 10 – 16. 2001.

[7] L. A. Geddes, L. E.Beker, et. Al. Optimum electrolytic chloriding of silver electrodes. Medical & biological Engineering. Vol. 7. Pp. 49 – 56. 1969.

[8] A. Mayer-Gindner, A. Lek-Uthai, O.Abdallah, and A.Bolz. Newly explored electrical properties of normal skin and special skin sites. Biomedizinische Technik. Vol. 49. Pp. 117 – 124. 2004.

[9] Tatsuma Yamamoto et al. Measurement of low-resistance points on the skin by dry roller electrodes. IEEE transactions on biomedical engineering. Vol. 35. Pp. 203 – 209. 1988.