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A multichannel portable ECG system with capacitive sensors

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Abstract

Capacitive sensors can be employed for measuring the electrocardiogram of a human heart without electric contact with the skin. This configuration avoids contact problems experienced by conventional electrocardiography. In our studies, we integrated these capacitive electrocardiogram electrodes in a 15-sensor array and combined this array with a tablet personal computer. By placing the system on the patient's body, we can measure a 15-channel electrocardiogram even through clothes and without any preparation. The goal of this development is to provide a new diagnostic tool that offers the user a reproducible, easy access to a fast and spatially resolved diagnostic 'heart view'.

Keywords: capacitive ECG, non-contact measurement, body surface potential mapping

(Some figures in this article are in colour only in the electronic version)

1. Introduction

Monitoring the health state of the heart is one of the most important diagnostic methods in medicine. The electrocardiogram (ECG) is the primary diagnostic tool in cardiology, where the measurement of the surface voltage on standard positions is normally based on a galvanic contact of the ECG electrodes with the skin. Ag/AgCl electrodes are used in combination with an electrolyte to optimize the impedance between the electrode and the skin. Furthermore, suction electrodes can be positioned to avoid the application of an electrolyte and to improve the mechanical contact with the body.

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The voltage differences between well-defined electrode positions provide the standard leads according to Goldberger, Einthoven and Wilson (Bronzino 1999) . In contrast to this established procedure, capacitive electrodes measure the surface potential without galvanic contact with the skin (Lopez and Richardson 1969, Prance *et al* 2000). This method eliminates the need for an electrolyte, significantly simplifying the handling of electrodes. A capacitive ECG measurement can be accurately obtained through several layers of clothing, removing the need for direct skin contact by the electrodes (Lim *et al* 2007, Ueno *et al* 2007). This qualifies capacitive systems for long-term ECG measurements. A comparison of such capacitive coupled electrodes with Ag/AgCl electrodes and electrodes with a conductive foam layer has been demonstrated (Gruetzmann *et al* 2007). Another way to measure the health state of the heart without direct contact with the patient is magnetocardiography (MCG) based on superconducting quantum interference devices (SQUIDs) (Schilling *et al* 1996, Barthelmess *et al* 2001). However, sensor cooling with liquid nitrogen or helium is required and the system costs are orders of magnitude higher. By body surface potential mapping (BSPM) spatial resolution of the electric potentials on the body surface with conventional galvanic electrodes is possible, but the exact position of each electrode relative to the others and relative to the body has to be documented precisely for reproducible measurements. Especially in the field of acute care this BSPM provides additional information about the heart state in comparison to the standard 12-lead ECG (Finlay *et al* 2005). BSPM is used to localize cardiac abnormalities like the early diagnosis of acute myocardial infarction (Carley 2003). For this surface potential mapping, the sensor positions in the electrode array have to be fixed. The arrangement of electrodes for galvanic contact in a mechanically fixed array is difficult, since contact strength must be adjusted for individual electrodes. With capacitive electrodes these problems can be addressed more easily, especially for BSPM measurement (Clippingdale *et al* 1994). Other capacitive electrode systems have been demonstrated by Harland *et al* (2005) and Lim *et al* (2006). Capacitive single electrode systems, which allow the measurement of the heart rate (Lee *et al* 2005), are available in the market. Our capacitive ECG system provides a measurement array including 15 capacitive electrodes, measuring and displaying the ECG in a standard time plot or as BSPM even through clothes by placing the system on the subject's chest. Therefore, we integrated the sensor electrodes in a compact housing and mounted them below a Tablet PC, including the analogue–digital (A/D) converters and the software for filtering and display purposes. This mobile system differs from the known capacitive ECG systems (Harland *et al* 2005, Lim *et al* 2006) by the flexible mounting of the electrodes and the integration of all components in one portable system. This 15-channel capacitive ECG (cECG) system measures signals of the same quality for medical diagnoses as conventional ECG systems (Oehler *et al* 2006, Oehler and Schilling 2007). In figure 1, the complete ECG system is depicted. It consists of a Tablet PC with 15 electrodes including a reference channel attached to a flat housing below the PC case. Also, the option to use an external capacitive electrode as the reference channel exists. The data acquisition is accomplished by a 16-channel 16 bit A/D converter card in the Tablet PC together with LabWindows/CVI (National Instruments) based data acquisition software. The software also has the capability for real-time noise filtering.

2. Experimental setup

2.1. Sensor

The capacitive electrodes used in our system are based on a metallic electrode plate, which is insulated from the skin by a thin plastic film (Mueller *et al* 2007). This geometry is similar

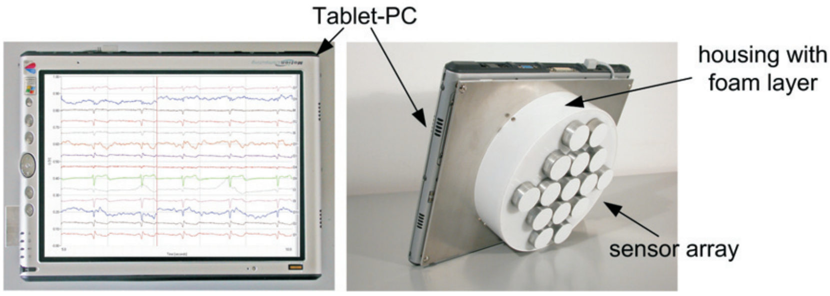

Figure 1. Tablet-PC-based multichannel ECG.

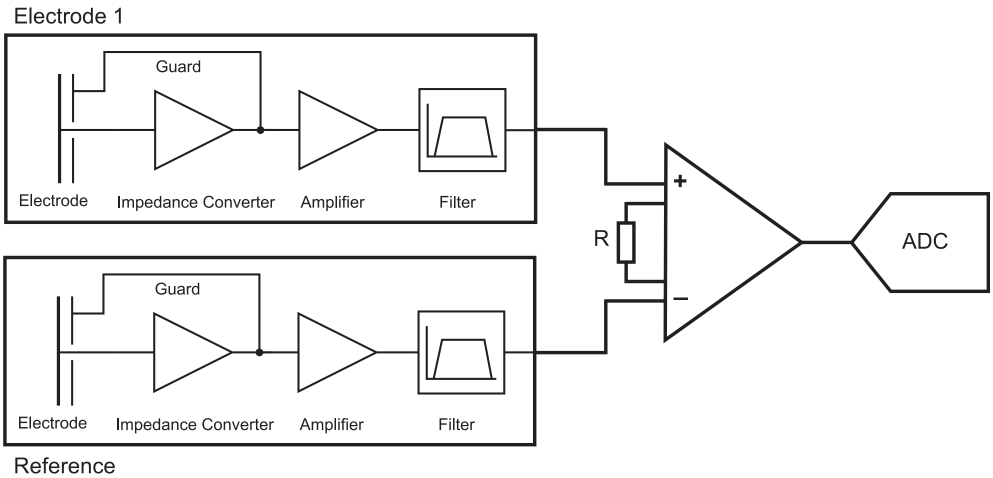

Figure 2. Sensor block diagram in differential configuration.

to the electrodes used by Lim *et al* (2006). On top of the electrode plate there is the signal-processing electronics containing the impedance converter, an amplifier and a filter circuit. For shielding purposes, the electrode and the electronic circuit are integrated in a metallic case to prevent the influence of disruptive external electric fields. Additionally, the electrode plate is shielded by an internal guard circuit surrounded by shielding layers, effectively shielding static charges. Power line interferences in the cables can be reduced through the application of noise suppression filters in the Tablet PC. The software also removes baseline wander and high-frequency disturbances. The power supply of the electrodes (± 5 V) is from a dc/dc converter connected to the USB port of the Tablet PC. Figure 2 illustrates the capacitive electrode's setup including the shielding configuration, the signal processing unit and the differential amplifier.

The electrode plate has a diameter of 26 mm and the whole sensor has a height of 15 mm. An LMC6084 operational amplifier from National Semiconductor is used for impedance conversion, preamplification and bandpass filtering. This circuit provides an extremely high input impedance of above 10^{15} Ω. The integrated preamplifier has a gain factor of 2. After amplification, the signal is bandpass filtered from 200 mHz to 80 Hz (second-order Butterworth). The complete electronics is included in the sensor. Figure 3(a)

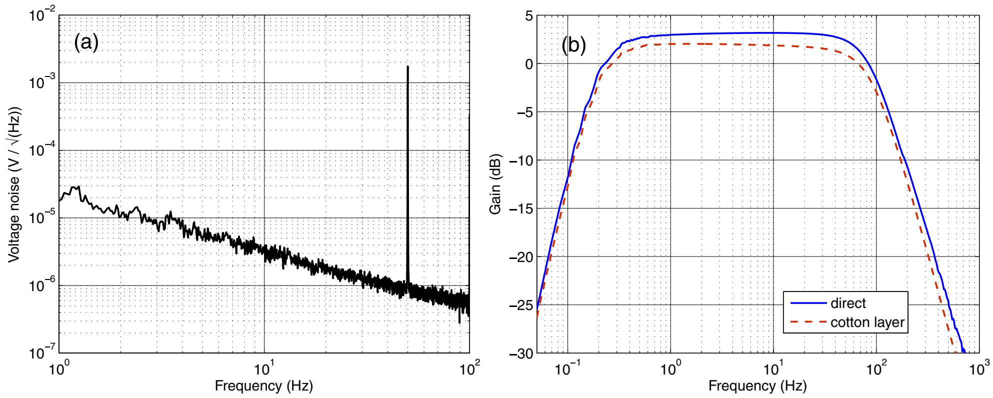

Figure 3. Voltage noise spectrum and frequency response of the capacitive sensor.

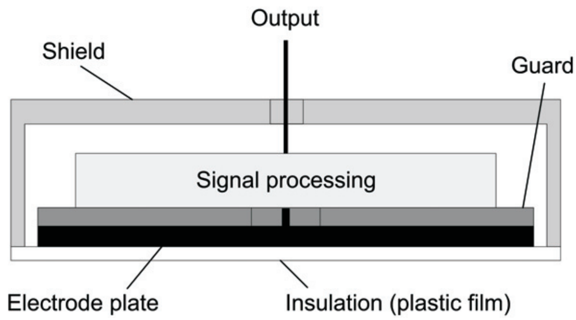

Figure 4. Sensor configuration (diameter: 30 mm, height: 15 mm).

shows the voltage noise spectrum of the sensor placed direct on the skin. The sensor has a noise level of about 5 μV Hz$^{-1/2}$ at 10 Hz. Figure 3(b) shows the change of the frequency response due to the existence of a cotton layer (0.4 mm) between the electrode and the body. Only a small decrease of the gain with increasing frequency can be observed. The decrease of the gain in comparison to the direct contact is caused by the increased distance between the electrode and the subject by the cotton layer thickness.

Figure 4 provides a cross-sectional view of the electrode. Within the electrode case are the aluminium housing, the signal processing unit, the guard circuit and the electrode plate. The bottom of the electrode is covered with a plastic film.

2.2. System

The 15 electrodes are attached to a fixed sensor array with a mean distance of 33 mm between the centre of the sensors. The electrodes are adapted to the contours of the patient's body surface by a foam layer behind the sensors. There is a 7 mm spring travel behind each sensor to enable proper contour adaptation when the system is placed at different positions on a patient's body. The sensors are used in a differential configuration with high common-mode rejection instrumentation amplifiers INA126 to further reduce power line interferences. The user has the option to select one electrode from the fixed sensor array as the reference channel

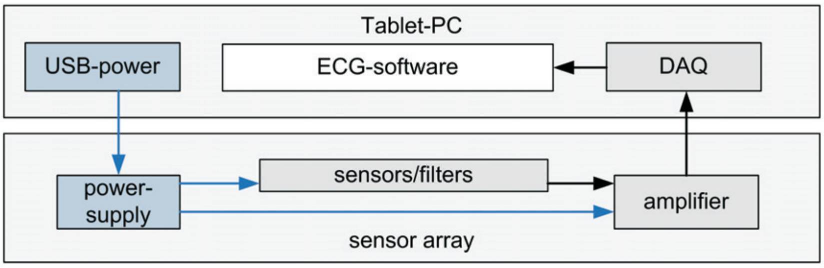

Figure 5. Signal flow and power distribution.

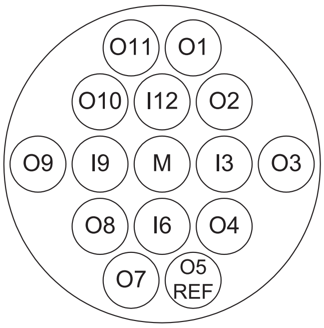

Figure 6. Sensor arrangement of the array (view to the body).

to all other electrodes by electrically connecting this electrode to the reference input on the amplifier board. An external reference can also be employed. Grounding is achieved only for this experiment via patient contact with the system plate or via an external cable. The amplifiers are configured for a gain factor of 50. The outputs of the amplifiers are sampled at 500 Hz by a National Instruments NI-6036 data acquisition card. An overview of the signal flow is illustrated in figure 5.

The Tablet PC is based on a 1.1 GHz processor and includes 512 MB RAM and a 12 TFT display. The power consumption of the sensor array is about 500 mW, so the battery time mainly depends on the Tablet PC power consumption. USB power is conditioned by a dc/dc converter to generate a symmetric supply of power for the amplifiers and the electrodes. This power, provided by the Tablet PC's built-in battery, is sufficient for successful ECG measurement for about 1 h. The control and display software is based on LabWindows/CVI 8 from National Instruments. It provides a real-time view of the measured cECG on the Tablet PC screen. After conversion of the input data, the software applies digital filters to remove noise and calculates the 2D plots of the selected sensors in a predefined arrangement. Figure 6 illustrates the sensor arrangement of the system as provided on the Tablet PC screen. Sensor O3 is on the patient's left side, O9 on the right, and the reference electrode (REF) is placed in the direction of the patient's left leg. The nomenclature of the sensors follows a clockwise numbering in inner and outer circles around the middle electrode M. The selected arrangement realizes a close coverage of the heart area (even outer channels, I3-I12,M) with inclusion of electrodes as far as possible from the heart (odd outer channels). To obtain a cECG interpretation corresponding to the standard ECG leads, it is preferable to place the reference electrode close to the right leg. This can be achieved with an external reference sensor.

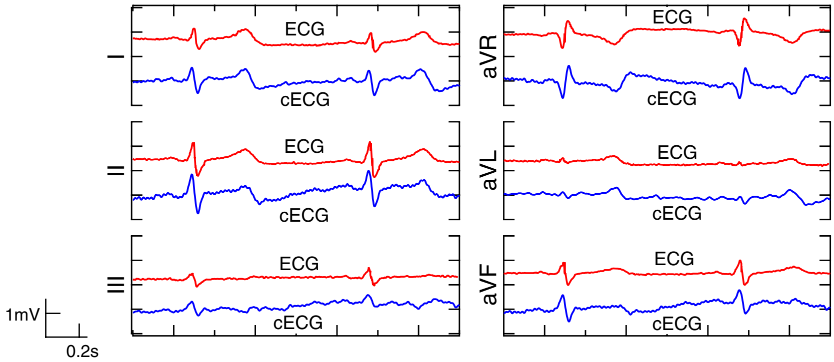

Figure 7. Comparison of ECG versus cECG of the same patient at the same time.

In our experiments, one of the internal fixed-array sensors was selected as reference, demonstrating cECG measurement capability in this compact configuration. The overall diameter of the sensor array is 185 mm, optimized to ensure that all sensors can easily adapt to the patient's body surface.

2.3. Diagnostic procedure

All cECG measurements except the comparison (figure 7) were performed through a cotton shirt on the patient's chest. No preparation was necessary for the actual measurement: the system was placed on the patient's chest and the data acquisition was started. At the start of the measurement, the subject was lying on a bed, breathing normally and holding the system on his chest. Grounding was achieved by the patient holding the aluminium plate at the bottom of the Tablet PC. For the measurements without grounding, the system was placed on the patient's chest by another person. Measurements were taken on all 14 channels simultaneously for a time $t = 60$ s. Electrode O5 was selected as the reference channel for all experiments with the sensor array. The data presented here were taken from the left chest in the region above the heart.

3. Measurement results

3.1. Comparison with conventional ECG

To directly compare the measured signals of the cECG with conventional Ag/AgCl-based electrodes, the capacitive electrodes were used separated from the Tablet PC array. The capacitive and the conventional sensors were pair-wise placed close to each other at the standard lead positions, one at the right and left shoulders, one on the left hips and the reference at the right hips (Einthoven/Goldberger). Figure 7 shows the simultaneously measured signals for the ECG and cECG.

3.2. Time trace

Standard ECG is typically presented in a graphical format as voltage versus time. Figure 8 shows the time traces of the 14 channels of the sensor array (figure 6) with QRS detection. The signals were only bandpass filtered by the integrated filter circuit with 0.2–80 Hz bandwidth in the sensors. This time trace view is updated in real time on the display of the Tablet PC

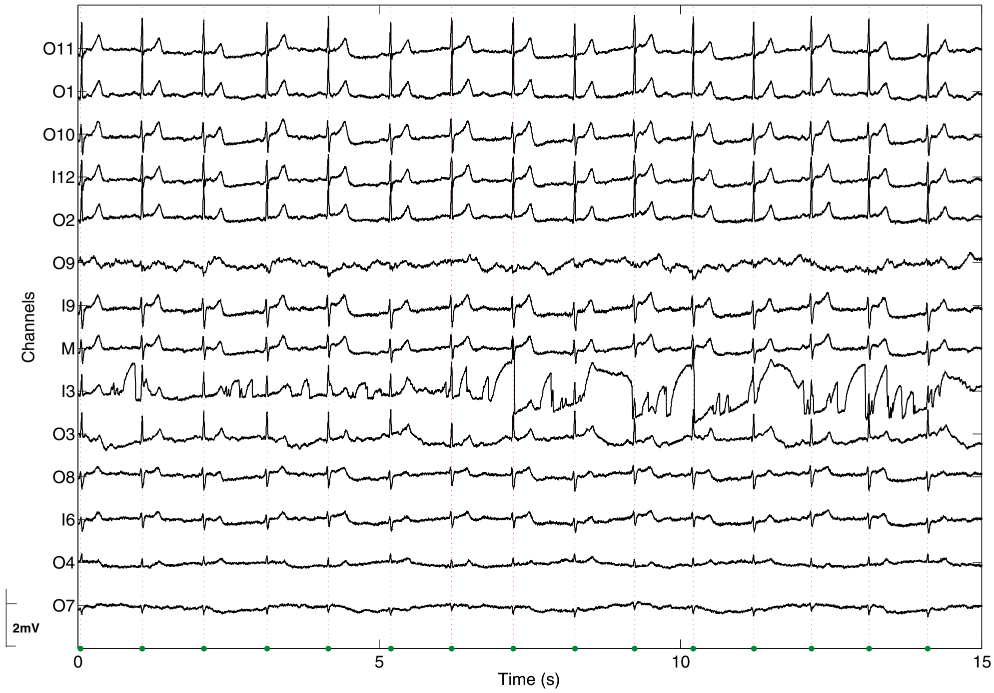

Figure 8. cECG time trace of all channels recorded over 15 s with QRS detection.

during the cECG measurement. The channel I3 in figure 8 displays some artefacts especially during the 7–15 s period. This was intentionally chosen to demonstrate how the system reacts to weak coupling. In such instances, the system can simply be geometrically readjusted to obtain a stable signal in all channels.

3.3. 2D plot

One of the main advantages of this system is the laterally fixed position of the sensors in the array, so the lateral distance between the measured potentials on the body is well defined, despite the elastic adaption to the body surface. Figure 9 shows the QRS complexes with T-wave of the channels with their relative position on the body. The measurement was taken on the left chest of the body and the coordinates represent the view of the body surface. The channel on the right side in the lowest line had been selected as reference and is represented by REF.

Figure 10(a) shows an interpolated contour plot of the cECG at the R-wave position in time. All ECG channels are plotted in a butterfly plot (figure 10(b)). The 80 ms line represents the time point of the potential plot. The black squares in the contour plot correspond to the sensor positions of the array on the chest. The difference between contour lines is 0.1 mV. The colour scale covers the range of the plot in figure 10(b). A dipole behaviour can be observed between the area on the upper right side (patient's left side) around electrodes O1–O3 and the area on the lower left side (patient's right side) around electrodes O8 and I9. The analysed heart dipole direction does not change relative to the body when the sensor array is rotated relative to the chest. So the displayed dipole direction is independent of the position of the reference electrode.

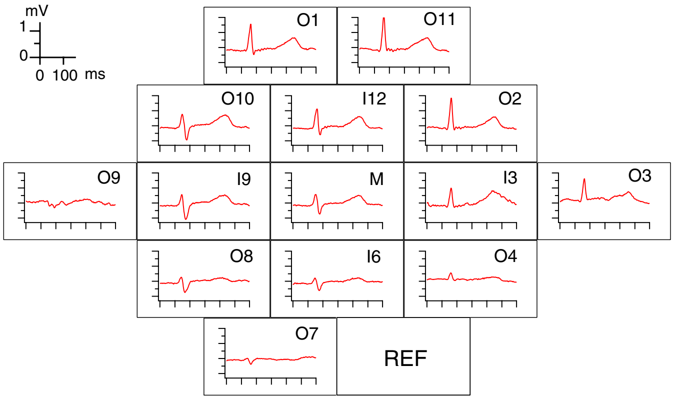

Figure 9. cECG at the positions of the sensor array.

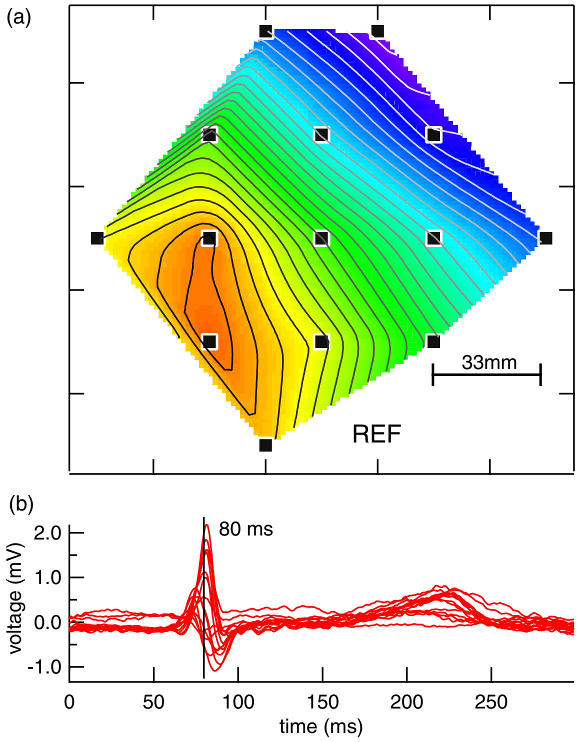

Figure 10. Contour plot (a) and butterfly plot (b) of the capacitive ECG.

3.4. Grounding

The non-contact capability of the capacitive system enables ECG measurements without direct skin contact. This system does not require the normal reference ground potential due to the differential sensor configuration. Thus, ECG measurements are possible without direct ground contact. Figure 11 illustrates the difference between ECGs recorded with (figure 11(b)) and without (figure 11(a)) grounding the patient and with software filtering (figure 11(c)). In the

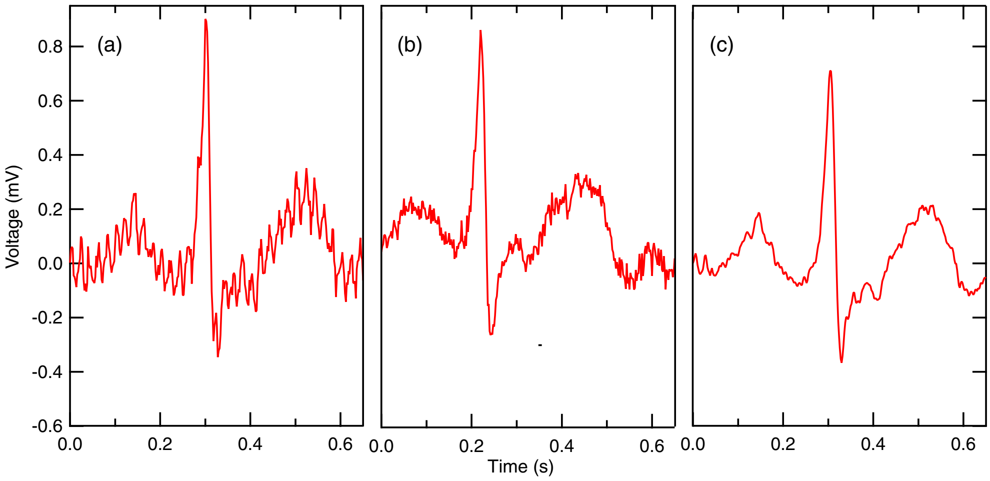

Figure 11. ECG measurement without ground connection (a), with ground connection (b) and with software filters (c).

non-grounding configuration, only the power-line noise increases, but the amplitude is quite low. In a noisy environment, an additional notch filter can be easily applied for software filtering (in figure 11(c), a second-order bandpass filter with a bandwidth of 0.5–40 Hz in addition to a 50 Hz notch filter has been applied to the grounded signal). The choice of the filter parameter was made similar to other groups (Lim *et al* 2006). The measurements shown in figures 11(a) and (b) only employed the built-in filter of the sensor; no extra digital filters were employed.

4. Discussion

The 2D plot of the surface ECG illustrated in figure 9 shows an expected decreasing behaviour of the R-peak height in direction to the reference electrode. The QRS complex changes in this direction. In the contour plot the expected dipole behaviour does not point to the direction of the reference channel, confirming that the potential map reflects the ECG potentials of the patient. Additional measurements have confirmed the independence of the dipole direction of the angle of the sensor array enabling a robust determination of the heart vector.

Nevertheless, the shape of the ECG signal depends on the position of the array on the chest and especially on the chosen reference position. For reconstruction of the standard ECG, the optimum position of this capacitive reference channel is the subject of further studies. The spatial resolution of the system is limited by the dimension of the electrodes. Decreasing the electrode dimension also reduces the signal quality of the ECG (Ueno *et al* 2007). The sensor arrangement shown and the size of the sensor array are best adapted for adult subjects, so for diagnostics in the case of acute myocardial infarction, the system is currently clinically under test. For infant ECG measurements, only a part of the sensor array is used for ECG measurement. For the diagnostic use in cases of emergency, the adaption to the body has to be optimized. Capacitive electrodes are very sensitive to motion artefacts, especially in the case of a non-fixed configuration. The system effectively reduces the artefacts by breathing or pulse movements through the elastic suspension of the sensors. Grounding the patient only affects power-line noise amplitude in the signal. For increased usability of the ECG system, measurements can be taken without an external galvanic grounding connection to the patient. When used without external grounding, the system is qualified to offer a real non-contact measurement. Even when a ground connection is used over an external wire in a noisy environment, the setup time is significantly lower than that of a traditional galvanic system, especially for a high number of channels. The calculation of the contour plot in combination with the non-contact electrodes provides a very fast and easy mapping of the health state of the heart. This view offers a 'window to the heart' by positioning the system on the chest and watching the heart beating with spatial resolution of the electric potential distribution.

5. Conclusion

We described a system based on capacitive electrodes for measuring a multichannel ECG with a fixed sensor array. The integration of the sensor array in a Tablet PC allows a very compact affordable ECG system especially for easy access to the measurement of body surface potential maps. The measurements were taken through clothes. No ground contact is required to measure a multichannel ECG. The Tablet PC provides a new, fast diagnostic tool through the real-time view of the electrocardiogram without any preparation procedure.

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