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Ultra-Wearable Capacitive Coupled and Common Electrode-Free ECG Monitoring System

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***Abstract*—Nowadays, transfer of the health care from ambulance to patient's home needs higher demand on patient's mobility, comfort and acceptance of the system. Therefore, the goal of this study is to proof the concept of a system which is ultra-wearable, less constraining and more suitable for long term measurements than conventional ECG monitoring systems which use conductive electrolytic gels for low impedance electrical contact with skin. The developed system is based on isolated capacitive coupled electrodes without any galvanic contact to patient's body and does not require the common right leg electrode. Measurements performed under real conditions show that it is possible to acquire well known ECG waveforms without the common electrode when the patient is sitting and even during walking. Results of the validation process demonstrate that the system performance is comparable to the conventional ECG system while the wearability is increased.**

I. Introduction

Monitoring of heart activity using ECG is one of the most important and commonly used medical investigations. Long term heart activity recording in patient's home allows to determine and predict the development of patient's cardiovascular system health and to diagnose chronic heart diseases [1]. The commonly used ECG monitoring systems capture signals from body surface using exactly located wet electrodes on patient's body, where conductive gels are used for an improvement of the measurement. Their application is constraining, time consuming and could cause allergic reactions. However, nowadays exist several types of dry electrodes the galvanic contact to the skin is still required. The new developed unconstrained method is based on the isolated capacitive electrodes which work without any galvanic contact to the skin and electrolytic gels are not necessary. Furthermore, methods used nowadays measure potential changes between two points on chest and one galvanic or capacitive coupled common electrode used as reference ground connected to the right leg [2][3][4]. The biggest challenge in capacitive ECG (cECG) sensing is high level of noise induced from power lines which has higher influence on cECG systems compared to conventional ECG systems due to capacitive coupling and the missing common right leg electrode.

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In this study the active electrodes with passive filtering in input amplifier stage were used to improve reduction of common mode signals which also allow to use a higher gain for better pick up of weak ECG signals caused by capacitive coupling.

II. Materials and methods

The developed device is based on the capacitive sensing principle. Figure 1 shows a short overview of the developed system. Signal obtained by isolated capacitive sensors placed on elastic belt is amplified by active front-end with ultra-high impedance amplifier. Output signals are fed to the analog processing board with an instrumentation amplifier and filters followed by A/D conversion and software signal processing.

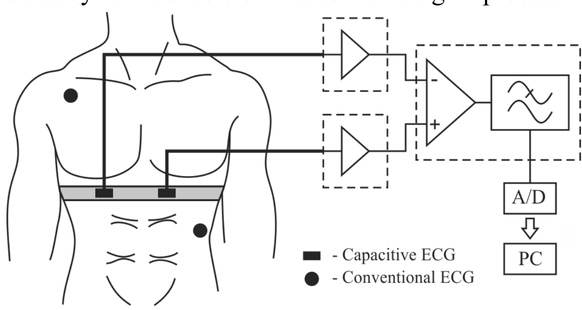

Figure 1. Functional diagram of developed system

A. Active Sensing Part

Input of sensing part is realized using insulated electrodes which forms a coupling capacitor (C_E) that carries bio-electric current from the body surface to the input amplifier. For experimental purposes due to simplicity one sided copper PCB with dimensions 5 cm x 3 cm is used. For the separation of sensors from patient's body as insulator an adhesive tape with thickness 35-50 μm is applied (see Fig. 4a). The capacitance of the electrode (C_E) depends mainly on the electrode area A, the relative permittivity ε_r and the thickness d of the insulator (1) and varies typically in tens of pF.

$$C_E = \varepsilon_0 \varepsilon_r \frac{A}{d}. \quad (1)$$

Alternating current from the electrode must be converted into voltage and the electrode's high input impedance has to be matched to the low impedance of subsequent circuits. This requirement is accomplished by ultra-high input impedance operational amplifier with FET input stage placed on the rear side of electrode (see Fig. 4b). Static charge of the capacitor C_E must be discharged over input bias path to ground and it can be done in different ways.

Biasing with resistors [5]: The resistance should be as high as possible to avoid degradation of input impedance (typically few GΩ), however equation (2) shows that such resistors bring high thermal noise

$$N_{therm} = \sqrt{4kTRB}, \qquad (2)$$

where k is Boltzmann's constant, T is absolute temperature, R is resistor value B is used bandwidth.

Biasing with diodes [6]: This method uses the leakage current of the diode in reverse polarization. Typically two anti-parallel connected diodes are used, however input resistance is not as high (typically hundreds of MΩ) as with resistor.

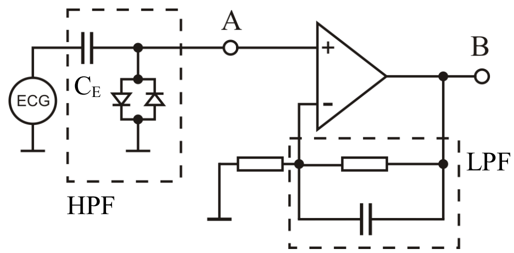

Figure 2. Circuit schematic of input amplifier

The ECG signal without a common electrode has very low amplitude and it is very difficult to separate it from noise created by the GΩ resistor. Furthermore, static charge is higher and discharging current over the GΩ resistor is not high enough and the amplifier is easily leaded out of its operation range. Moreover, such probes recover slowly from disturbances caused by motion artifacts such as breathing, walking etc. Therefore, the biasing with diodes is used. Table 1 illustrates the summary of bias circuits' properties.

TABLE I. BIAS CIRCUITS SUMMARY

Bias Type	Impedance	Cutoff frequency	Thermal noise	Recovery from artifacts
Resistor	High	mHz	High	Slow
Diode	Medium	Hz	Low	Fast

Input capacitance (C_E) in combination with bias resistance (R_{BIAS}) form the high pass (HP) filter (see Fig. 2). For input capacitance of tens of pF and bias resistance of hundreds MΩ the cutoff frequency (f_C) ranges in few Hz (3).

$$f_C = \frac{1}{C_E \cdot R_{BIAS} \cdot 2\pi}. \qquad (3)$$

Higher cutoff frequency influences the useful ECG spectrum and the input ECG wave (see Fig. 3a) is derivative (see Fig. 3b). Therefore, the first order low pass (LP) filter in the feedback circuit of input amplifier is applied. The LP filter acts as an integrator and repairs the ECG wave (see Fig. 3c). Moreover, it protects the input stage from saturation caused by high power line noise. Thus, we could use higher gain (100x) for better pick up of weak signals. The Correct time constant of integrator and AF characteristics of LP filter were proved during system evaluation when both probes were fixed to a PCB which was connected to arbitrary generator.

Guarding and shielding methods, necessary for reducing leakage currents and noise from supply, are used during PCB layout design.

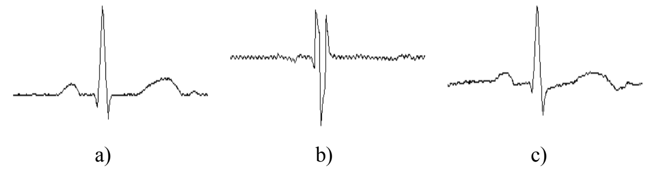

Figure 3. ECG waveform: a) Input waveform – arbitrary generator, b) Derivative – amplifier input (A), c) Integrated – amplifier output (B)

B. *Analog Pre-Processing Board*

The pre-amplified signals from each electrode are fed over twisted cables to reduce the common mode signals to the differential amplifier. However, CMRR of the instrumental amplifier is high (120 dB), tolerance in electrode impedance (Z_E) means different amplitudes and phases of common mode signals at the input of the differential amplifier what decreases the CMRR of the whole system. Equations (4) and (5) illustrate the differential voltage caused by asymmetry in Z_E proportional to common mode potentials (V_{CM}). To improve suppression of interferences the difference of impedance (ΔZ_E) must be minimal or V_{CM} reduced as much as possible [7]. In our case V_{CM} is reduced by the LP filter in the active electrode.

$$V_{diff} = V_{CM}\left(\frac{Z_{E1}}{Z_{E1}+Z_{amp}} - \frac{Z_{E2}}{Z_{E2}+Z_{amp}}\right), \qquad (4)$$

$$V_{diff} = V_{CM}\frac{\Delta Z_E}{Z_{amp}}, \qquad (5)$$

where Z_{amp} is amplifier input impedance.

Differential amplifier is followed by the sixth order LP Bessel filter with cutoff frequency of 150 Hz to remove spectral components above ECG useful spectrum. The board is realized on a separated PCB (see Fig. 4c).

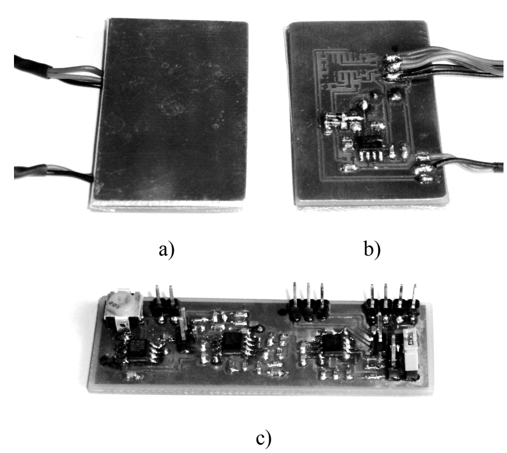

Figure 4. View of assembled system: a) Insulated electrode, b) Input amplifier on rear side, c) Analog pre-processing board

C. Digital Signal Processing

Output signal from the analog front-end after hardware processing is digitalized at 2 kS/s sampling rate by 24 bit A/D converter using data acquisition system (National Instruments), transferred to the PC and stored for signal processing using MATLAB.

The significant bandwidth of the ECG waveform is in range from DC to 40 Hz. Thus, sampled ECG signal was optimized by decimation (decimation factor 10) for reducing time consumption of filtering process. However, the sample frequency by decimation is decreased to 200 Hz, the Shannon-Kotelnik sampling theorem is still satisfied. Offset of base line is removed before filtering. Equation (6) illustrates basis of realized baseline removal process.

$$Y = Y_n - \bar{Y}_0 \; ; \; n = N - 1, \qquad (6)$$

where $\bar{Y}_0 = \frac{1}{N}\sum_{n=1}^{N} Y_n$, Y_n is actual sample of input signal with offset, Y is output vector with removed offset, N is number of samples.

Two 64 order FIR filters with Gaussian windows are used. Common mode noise is removed by a LP filter with cutoff frequency of 40 Hz and eventual motion artifacts (breathing, muscle activity) are suppressed with HP filter with cutoff frequency 0,4 Hz.

III. Results

Measurements were realized on a 25 years old healthy male to proof the developed concept under real conditions. Contact-less active electrodes were fixed on the subject's chest with an elastic belt without the presence of clothing. Figure 1 illustrates the measurement setup. Simultaneously recorded conventional ECG was used for validation purposes. During tests it was disconnected from subject's body to avoid any contact with system ground. To avoid galvanic contact with power line the battery power supply was used.

A. Motionless Measurement

The first experiment was realized while the subject was sitting. The obtained signal from our system is shown on Figure 5a. However the system was supplied with battery, the 50 Hz common-mode noise after differentiation by instrumental amplifier with CMRR up to 120 dB was still significantly influenced. The power line noise after applied signal processing was suppressed and well known QRS complex and T wave became visible (see Fig. 5b).

The placement of electrodes which was not according to the conventional ECG configuration and missing common electrode could be the reason for missing P wave. Recorded conventional ECG is shown on Fig. 5c for comparison.

B. Measurement with movement

The second experiment was realized while subject was walking. The purpose of this measurement was to test the influence of patient's movement on quality of measured ECG signal. Figure 6 shows the influence of walking. However, the influence of motion artifacts is significant, operating range was not exceed. To suppress those artifacts HP filters with different cutoff frequencies (in range 1 Hz to 10 Hz) were tested. Although, the T wave was destroyed, the QRS complex is visible. Results show that ECG was successfully obtained also during walking.

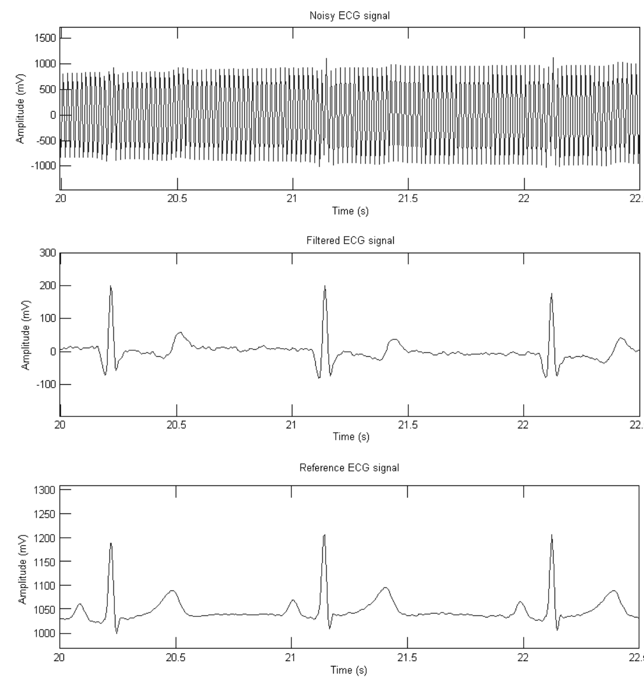

Figure 5. Measured motionless ECG: a) Output signal obtained from analog front-end, b) Filtered output cECG signal after signal processing with missing P wave, c) Simultaneously recorded conventional ECG

IV. Discussion

Today commonly used ECG sensors require galvanic contact of measuring probes directly to the skin and common electrode. According to the achieved results in laboratory conditions our research shows that it is possible to acquire ECG signal without right leg electrode in comparison to conventional ECG.

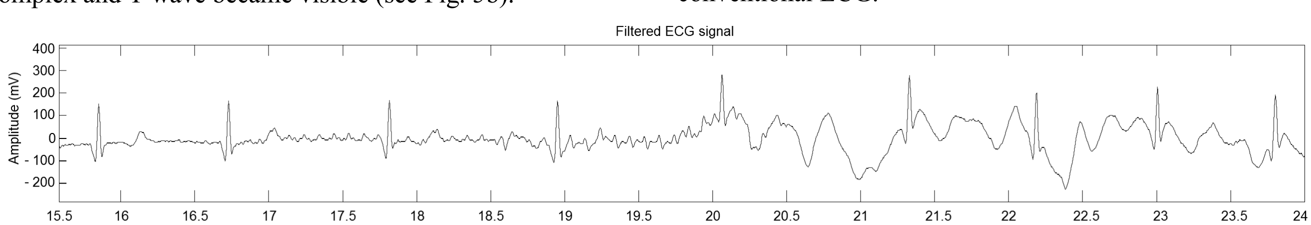

Figure 6. Measurement of ECG during transient from sitting to walking

For better signal quality the improvement of the system's CMRR could be helpful. Solution for mentioned purposes is future investigations in better fabrication of electrodes and shielding, where tolerances in electrode impedance have to be reduced to minimum. Also different materials, isolation surfaces, optimal electrode dimensions, capacities and shapes could be found out and tested. The common problem of ECG measurement is the influence of motion artifacts (caused by breathing, muscle activity and slow changes of impedance properties caused by body-sensor coupling) that could be reduced with better electrode fixation or new algorithms for the suppression of motion artifacts [8].

V. Conclusion

The developed concept is a promising approach for health care, because conventional ECG systems do not satisfy the requirements for long term measurements in home healthcare applications. The using of capacitive probes reduces the costs and application time (no gel is required), moreover electrodes after washing are reusable and no waste is produced. The results of the proposed system are comparable to conventional systems and the wearability of the system and patient's mobility were increased.

References

[1] I. Korhonen, J. Parkka, and M. V. Gils, "Health monitoring in the home of the future" IEEE Eng. Med. Biol. Mag., vol. 22, pp. 66–73, May/Jun. 2003.

[2] M. Ishijima, "Monitoring of electrocardiograms in bed without utilizing body surface electrodes" IEEE Trans. Biomed. Eng., vol. 40, pp. 593-4, 1993.

[3] Y. Lim, K. Kim and K. Park "ECG measurement on a chair without conductive contact" IEEE Trans. Biomed. Eng. vol. 53 pp. 956–9, 2006

[4] Y. Lim, K. Kim and K. Park "ECG recording on a bed during sleep without direct skin-contact" IEEE Trans. Biomed. Eng. vol. 54 pp.718–24, 2007

[5] D. Svard, A. Cichocki, A. Alvandpour, "Design and evaluation of a capacitively coupled sensor readout circuit, toward contact-less ECG and EEG", Biomedical Circuits and Systems Conference, 978-1-4244-7269-7, pp. 302 - 305, 2010

[6] M. Yu, S. Deiss and G. Cauwenberghs "Non-contact low power EEG/ECG electrode for high density wearable biopotential sensor networks" Proc. 6th Int. Workshop on Wearable and Implantable Body Sensor Networks pp. 246–50, 2009

[7] M. A. C. Rijn, A. Peper, C. A. Grimbergen, "High-quality recording of bioelectric events", In: Medical and Biological Engineering and Computing, Vol 28, No. 5, pp. 389-397, 1990

[8] S. Heuer, D. R. Martinez, S. Fuhrhop, J. Ottenbacher, "Motion artefact correction for capacitive ECG measurement", Biomedical Circuits and Systems Conference, pp. 113-116, 2009