J Ambient Intell Human Comput
DOI 10.1007/s12652-012-0153-8

ORIGINAL RESEARCH

Textile-based, contactless ECG monitoring for non-ICU clinical settings

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Received: 30 September 2011 / Accepted: 28 March 2012

Abstract Even though it might be beneficial to have continuous baseline data upfront, today's hospitalized patients are put on a vital signs monitor not earlier than they become bed-ridden. The aim of this study is to show the feasibility of a broader monitoring concept, embracing many more subjects, by monitoring contactless vital signs with sensors embedded in non-ICU settings. In order to conduct the experiments, a system was designed to measure electrocardiograms (ECG) unobtrusively by capacitive sensors in various patient fitments e.g., stretchers, hospital-beds and wheelchairs. Contact ECG and contactless ECG measurements by shielded textile electrodes are contrasted to each other to display the potential of contactless monitoring. Our experiments demonstrate that the contactless ECG recordings are of sufficient quality and later used for further standard ECG analysis like QRS-complex detection. We believe that our results are motivating for the potential future care facility featuring a personal, non-obtrusive and yet permanent health monitoring system.

Keywords Contactless electrocardiography · Clinical environments · Textile capacitive electrodes · Patient's freedom of movement · Future hospital

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1 Introduction

In today's budget-driven hospital environments, only critical patients are continuously monitored during their hospital time on bed. Subsequently, mobile patients are not monitored, even though it might provide useful baseline recordings for further diagnosis and treatment.

Besides budgetary restrictions, obtrusiveness is a stronger hindrance to permanent monitoring: no mobile patient is happy to push a cart with monitors along his way, just to maintain the optimal position of cable bound electrocardiogram (ECG) electrodes. Even when using a portable Holter-type device, electrode–skin contact still forms a delicate interface prone to degradation of conductive gel, motion artifacts and skin irritations. Metal electrodes in long term recordings might even lead to allergies and skin irritations and may result in pressure necroses (Aleksandrowicz and Leonhardt 2007).

Today's textile-based sensing technology shows a way towards unobtrusive recording of electrical biosignals by electrodes not in intimate contact with the skin but capacitively coupled only. Another important aspect of unobtrusiveness is the integration of such sensors in everyday clinical objects, thus avoiding any kind of wirings to a patient and still providing monitoring options.

Published online: 14 July 2012

1.1 Brief history of capacitive body sensing

The first capacitive recording of an ECG signal without conductive body contact was described by Richardson (1967). The surfaces of his capacitive electrodes were electrically insulated and remained stable for long-term monitoring. Wolfson and Neuman (1969) also designed a capacitive coupled Electrocardiograph using high input impedance amplifiers based on MOSFET. David and Portnoy (1972) used for the same purpose an integrated arrangement of an insulated electrode and an impedance matching configuration with a Fairchild μA740 operational amplifier enclosed in a plastic housing.

Betts and Brown (1976) built circular disc type conductive plastic electrodes in place of metal electrodes. They conducted a study on 200 patients while comparing his plastic with a wet pad system of electrodes. An integral unit of electrode and amplifier were designed by De Luca et al. (1979) to be used in the front end of a standard biopotential measurement. The module was enclosed in a metal shell, which worked as housing and ground contact as well.

Lately, there have been significant new developments in the subject of capacitive electrocardiography. Clippingdale et al. (1994) presented an array of ultra-high ohmic sensors for heart imaging at the University of Sussex. The array comprises 25 spring-mounted sensors, arranged in a five by five array, with 32 mm distance between sensors.

Prance et al. (2000) published their study on a potential probe for human body scanning. The probe was optimized for ultra-low noise with an instrumentation amplifier and had a bandwidth of 0.01–100 kHz to detect a wide range of electrical activity from the body. Further advancement in this probe with remote sensing of the human body was demonstrated by Harland et al. (2002). They presented the capability of sensing ECG signals from a distance of 1 m away from a clothed body with high resolution ECG.

The term *capacitively coupled non-contact electrode* (CCNE) was introduced by Lee et al. (2004) at the QUASAR research team. Their compact sensor was claimed to be able to record ECG signals without any galvanic contact through clothing.

Lim et al. (2007) presented a study on contactless ECG measurement on a bed during sleep. They implemented an array of 8 copper clad capacitive electrodes (4×4 cm^2) with embedded electronics and a large conductive textile electrode used as ground plane. To measure heart rate, the R-peak was detected from one of the 8 channels, sorted by its signal quality.

An interim study of capacitive ECG measurement in clinical practice was conducted by Eilebrecht et al. (2009) at the Aachen University Clinic. Their group integrated stiff capacitive electrodes into a pillow to be used in a clinical bed or a chair with active driven reference electrode, analogous to the "driven right leg circuit".

1.2 Textile-based capacitive electrocardiogram

Our group successfully integrated textile ECG electrodes into an automotive environment to capacitively record biosignals (Chamadiya et al. 2010, 2011), whereas in the following we propose a textile electrode based, contactless ECG monitoring system added to clinical settings. In this way, patient's contactless ECG monitoring can start immediately in the ambulance on a stretcher, during hospital stay in bed or during post-procedure mobility in a wheelchair (Fig. 1). The objective of our experiments was to acquire contactless ECG synchronous with conventional ECG and to compare their quality by implementing one of the standard ECG signal processing algorithms on QRS detection.

Unlike most of the earlier work on capacitive ECG, both measurement and reference electrodes have been implemented as textile electrodes within our study. Comparisons of heart rate and power spectral density of capacitive-coupled electrocardiogram (CCECG) and conventional electrocardiography with various clinical settings are also presented. We considered scenarios where effortless monitoring would be beneficial, like being poly-traumatized carried on a stretcher or lying in bed or using a wheelchair.

2 Theory of capacitive sensing

2.1 Capacitive-coupled ECG detection

Electrical potentials generated by the heart are conducted through the body's tissue and are measurable at the epidermis (outer skin layer). Equipotential lines of these

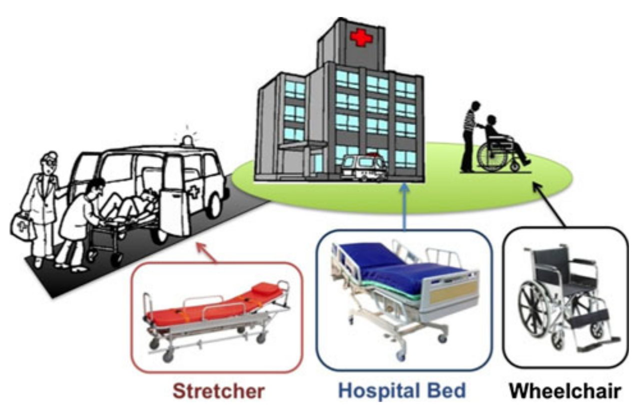


Fig. 1 An illustration of non-ICU clinical environments considered for the CCECG system

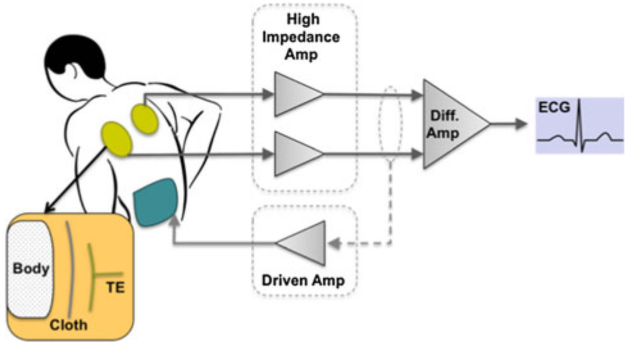


Fig. 2 A general block diagram of the CCECG system

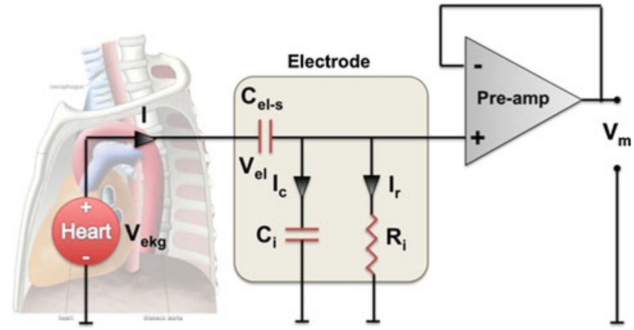


Fig. 3 Circuit network of the CCECG input stage

potentials create a body surface distribution of them to be mapped (Malmivuo and Plonsey 1995). Conventional electrocardiography uses a galvanic contact with the skin to acquire these potentials (Bronzino 2000). However, the body surface can be considered as conducting plate and if there is any conductive surface close to the body separated by an electrical insulation, they will form a capacitor (C_{el}) shown in Eq. (1) (Clippingdale et al. 1994). Here, clothes, air and any material between the two surfaces act as a dielectric material of the capacitor as shown in Fig. 2.

$$C_{el} = \varepsilon_r \varepsilon_0 \frac{A}{d} \tag{1}$$

Figure 2 sketches the basic configuration of a capacitive ECG system. The body and capacitive electrode (TE) form a capacitor (C_{el}) to couple the potential from the body to a high impedance amplifier (Clippingdale et al. 1994). The input amplifier works as a buffer to transfer the signal from a high impedance source to its low impedance output for the next stage of signal conditioning. Signals from the buffers are fed to differential amplifiers to minimize common mode signals and to amplify the differential signal.

Further improvement in the common mode cancellation is achieved by a driven seat circuit. The circuit uses the unity gain signals from the buffer to feed it back to the body through a driving amplifier to the driven seat electrode after adding and amplifying the signals (Kim et al. 2005; Lee et al. 2009).

2.2 High input impedance pre-amplifier

The capacitance formed between a body and the capacitive electrode is very low (Prance et al. 2000) and varies with separation distance (d) in Eq. (1) (approx. 1–100 pF), hence the resulting contact impedance remains in the range of giga ohms. The typical amplitude of a surface ECG signal lies in the range of 1–2 mV. The amplitude of the ECG signal is further reduced by coupling it capacitively to a measurement device. In this case, it is crucial to appropriately match sensor and amplifier impedances to acquire signals with minimum losses.

The input stage of the capacitive coupling is depicted in the Fig. 3. According to Kirchhoff's voltage law, the potential distribution in the loop is derived as follows.

$$V_{ekg} = V_{el} + V_m$$
$$V_{ekg} = \frac{1}{C_{el-s}} \int I\, dt + V_m$$
$$V_{ekg} = \frac{1}{C_{el-s}} \int (I_c + I_r)\, dt + V_m$$
$$V_{ekg} = \frac{C_i}{C_{el-s}} V_m + \frac{1}{R_i} \int \frac{V_m}{C_{el-s}}\, dt + V_m$$
$$V_m = V_{ekg} - \frac{C_i}{C_{el-s}} V_m + \frac{1}{R_i} \int \frac{V_m}{C_{el-s}}\, dt \tag{2}$$

In Eq. (2), C_{el-s}, R_i, and C_i represent the body-electrode capacitance C_{el} in series with another capacitor, total input resistance R_i and total input capacitance C_i respectively.

Here from the Eq. (2), it can be deduced that two conditions are necessary to minimize signal attenuation with $X_{C_{el-s}}$ being the impedance of C_{el-s}.

$$C_{el-s} \gg C_i \tag{3}$$

$$R_i \gg X_{C_{el-s}} \tag{4}$$

This means that the desired amplifier needs a smaller input capacitance than the skin–electrode capacitance and higher input impedance than the skin-contact impedance to optimize the signal coupling and minimize signal attenuation.

2.3 Parasitic capacitance and guarding

A potential difference between two wires lying adjacent to each other affects each other's electric field and thus both act like a capacitor (Spinelli and Haberman 2010). As depicted in Fig. 4a, a parasitic capacitor is generated in parallel to the input impedance of the preamplifier and hence undesirably reduces the overall impedance of the preamplifier.

To reduce this parasitic effect, guarding is commonly practiced as shown in Fig. 4b (Pallas-Areny and Webster

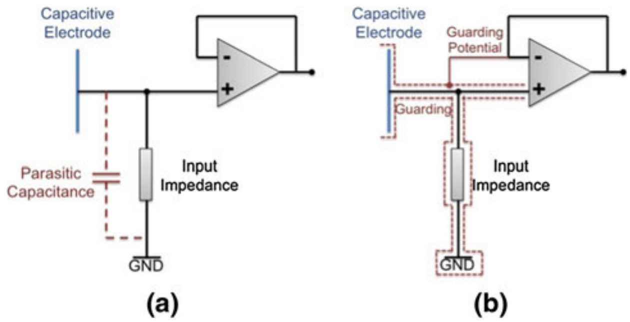


Fig. 4 Amplifier input-stage circuits showing **a** the parasitic capacitance, and **b** a guarding network as a solution to minimize parasitic capacitance

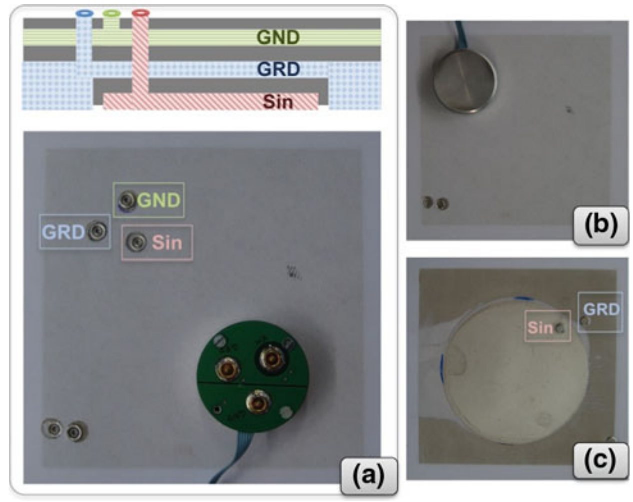


Fig. 5 Structure of the capacitive-coupled electrode; **a** rear view with open connections and different layers of the electrode (*top*), **b** rear-view with pre-amplifier plugged-in and **c** front view

1999). The guarding network surrounds the input channel and carries the same potential as the input. It tends to zero voltage difference between input wire and guarding network and hence no parasitic capacitance between them (Rich 1983).

2.4 ECG signal processing

Acquired ECG signals are usually fed into a computing platform for further signal processing. The objective of ECG signal processing aims to improve measurement accuracy and to extract information not readily available from the signal by visual analysis (Metting van Rijn et al. 1990). In this study, we aim to verify the quality of CCECG in comparison to conventional contact ECG recordings. Consequently, spectral analysis has been employed to detect noise components and other clinical relevant details.

The presence of a heartbeat and its occurrence in time is a very basic information required by all types of ECG analyses (Sörnmo and Laguna 2006), so we employed a well-known method to detect QRS complexes and the heart-rate (HR) from the CCECG recordings to compare it with conventionally acquired HR.

3 System components

3.1 Capacitive-coupled electrode

In our previous work (Chamadiya et al. 2008), we developed and experimented with stiff capacitor plates for ECG detection. Here we are presenting the design of flexible, textile-based capacitive electrodes. The electrodes were printed and developed with help of Institut für Spezialtextilien und flexible Materialien (TITV), Greiz, Germany. The electrode contains three conductive (silver printed) and three isolating textile layers (PES-Knitted substrate-7058, Thorey). A silver gel was used as electrode print and consisted of 75 % silver solvated in 2-butoxyethanol. The structure shown in Fig. 5a presents various layers of the electrode; sensing layer (Sin), guarding layer (GRD) and grounding layer (GND).

Guarding and grounding layers work as active and passive shields respectively for the sensor layer. As described earlier in Sect. 2.2, the guarding technique actively shields a surface against any parasitic effect (e.g., coupling with surrounding noise) by driving the guarding layer with the common mode voltage (Pallas-Areny and Webster 1999). Ground layer on top of the guarding layer shields the input further (Rich 1983). Each layer of the electrode is connected with the preamplifier through a conductive snap fastener as shown in the implemented electrode (Fig. 5a, b).

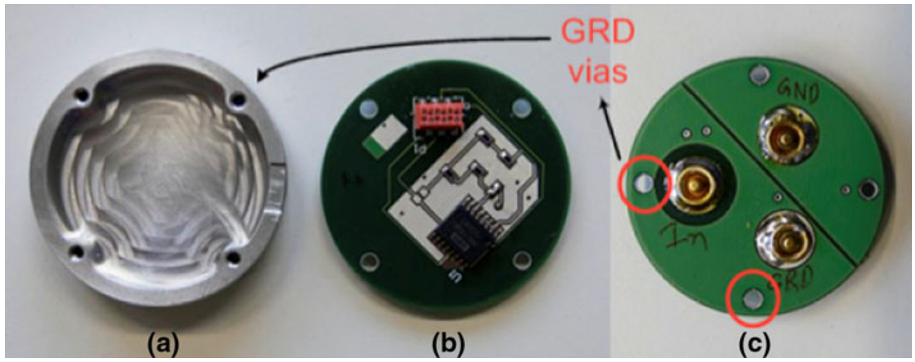
3.2 CCECG preamplifier

A high input impedance amplifier with biasing was needed to couple the ECG signal as explained in the Sect. 2.2. An ultra-low input bias current instrumentation amplifier INA116 (Burr Brown Corporation, USA) has been selected for this application. It features a very high input resistance (10^{15} Ω) combined with a very low input capacitance (0.2 pF) (Burr-Brown Corporation 1995), both are desired specifications for a preamplifier for capacitively-coupled bio-potential detection. A modular design containing the preamplifier (INA116) printed circuit board (PCB) and an actively shielded aluminum case is shown in Fig. 6.

A four-layer PCB was designed to shield the input signal from the textile electrode through conductive snap fasteners. Guarding and ground layers were also laid out on the bottom-connecting layer of the PCB, as shown in Fig. 6.

Two holes (left and bottom in Fig. 6c) are built as conducting vias and are used for holding the aluminum

Fig. 6 Pre-amplifier module: **a** module housing, **b** PCB layer of the electronics circuit and **c** PCB layer with snap connections to the electrode



housing with screws. This connection is added to the guarding plane on the top layer of the PCB (silver colored) ultimately joining the guarding pins on the INA116; and the guarding layer of the textile electrode through the snap fastener as well. Driving the housing with the common mode voltage guards the input by avoiding coupling to surroundings as mentioned earlier.

3.3 Signal conditioning

A capacitive ECG system is highly prone to noise due to its ultra high and sometimes fluctuating input impedance. In this case, signal conditioning plays a very important role to remove noise.

Capacitive-coupled signals from each of the CCECG electrodes are fed to a differential filter in an analog signal processing toolbox (90IPB, Frequency devices). The differential signals from the two CCECG electrodes are filtered with a window of 1–37 Hz by an 8th order elliptical band-pass filter. The signal is amplified before and after the filtering by a gain of 10 and 20 respectively.

The filtered signal is acquired by a LabVIEW (National Instruments, USA) program for further processing after digitizing it with a sampling rate of 5 kHz and a resolution of 16 bit with a data acquisition (DAQ) card (NI U-9162, National Instruments, USA). Here, the signal is digitally filtered by a 4th order Butterworth band-pass filter with a bandwidth of 1–40 Hz and a 4th order Butterworth notch filter with 50 Hz center frequency. The resulting signal is displayed by the LabVIEW application. As shown in Fig. 7, the regular ECG (R-ECG) signal captured by contact textile-electrodes (TITV-Greiz, Germany) is pre-processed by a bio-potential amplifier chip, RHA1016 (Intantech LLC, USA) described in (Mankodiya et al. 2010) and is finally fed to the DAQ card for parallel display and processing together with CCECG signal.

3.4 ECG signal post-processing

Stored R-ECG and CCECG signals are offline post-processed and analyzed within the MATLAB environment

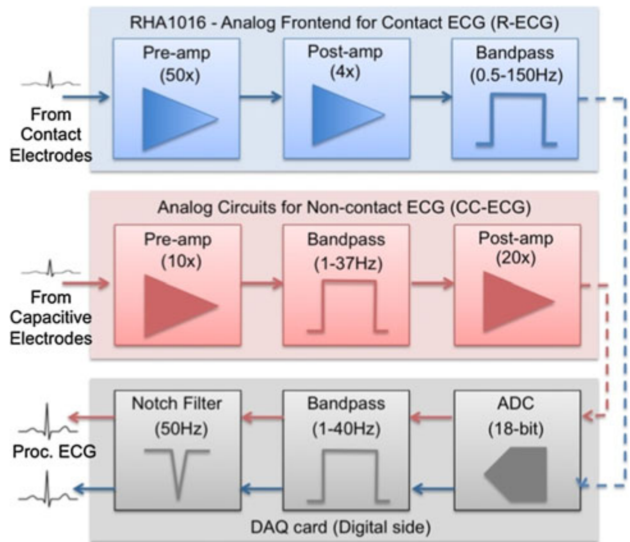


Fig. 7 Block diagram of regular contact R-ECG and CCECG signal conditioning

(MathWorks Inc., USA). The power spectral density of the R-ECG and CCECG signals were calculated and displayed and the routine "nqrsdetect" (The BioSig Project 2011) was adapted to detect QRS-complexes and their timing.

The *nqrsdetect* is an open source multi-rate digital signal-processing algorithm to detect QRS complex in ECG signals (The BioSig Project 2011). The algorithm incorporates a filter bank and decomposes the ECG signal into sub-bands with uniform frequency bandwidth (Afonso et al. 1999). Our adjusted routine calculates the RR interval timings and hence the heart-rate from the ECG signal under test.

4 CCECG set-up in clinical environments

In order to achieve the objective of unobtrusive monitoring, we equipped three objects frequently used by patients with a CCECG system: a stretcher, a bed and a wheelchair. Objects were chosen such that the patient's ECG can be monitored from the moment, they require urgent medical help until they leave the hospital.

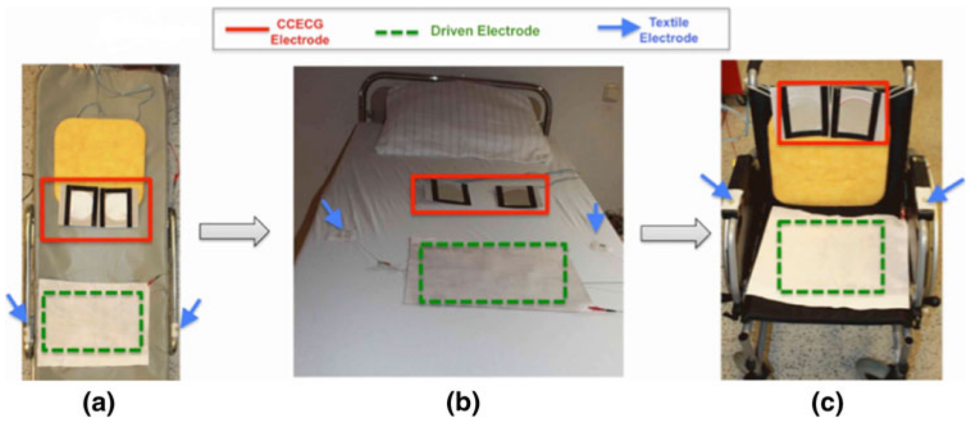


Fig. 8 Experimental setup for various clinical environments; **a** stretcher, **b** hospital bed and **c** manual wheelchair

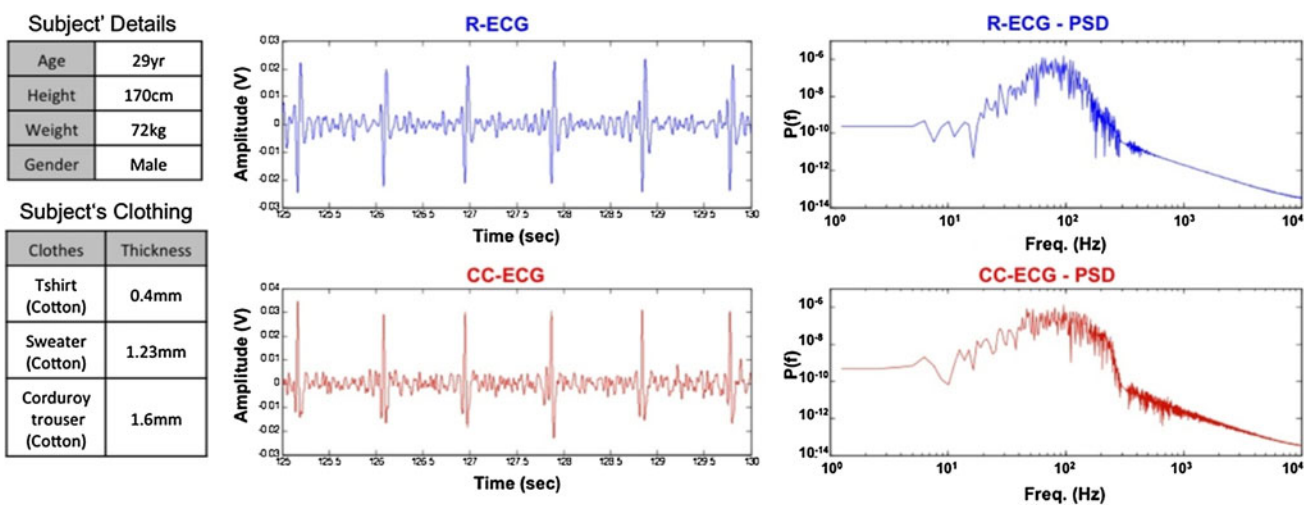


Fig. 9 ECG measurements on the stretcher for 5 s

4.1 Stretcher

A stretcher often is a patient's first-point of contact with the clinical environment. For us it is the first contact with the patient's back and hence furnishes a good platform to capture CCECG in ambulatory conditions. As shown in Fig. 8a, CCECG electrodes are simply placed in the thoracic section of the stretcher and are facing towards the back of the patient. No gelled electrodes have to be fixed on the patient. A square shaped driven electrode is located at the lumbar section of the stretcher. For validation purposes, R-ECG was taken by attaching conductive textile electrodes on the stretcher's hand-rest, to be easily reached by the patient's hands.

4.2 Hospital-bed

After the patient is admitted, they usually spend a major portion of their hospitalized time on a clinical bed. Figure 8b shows the locations of two CCECG electrodes on the bed-sheet. The contact ECG electrodes are also placed on both sides of the bed-sheet and are easily accessible by the volunteers' hands.

4.3 Wheelchair

During recovery phases in hospitals, patients are allowed to mobilize themselves on wheelchairs. Wheelchairs also provide firm contacts between the patient's back and the wheelchair's back-rest. On the other hand, a rigid contact of the patient's bottom within the seat facilitates a good driven electrode contact. Our wheelchair (Fig. 8c) was prepared for CCECG measurement by placing electrodes in the back-rest and the driven electrode on the seat. The contact ECG electrodes were fixed onto both arm rests.

5 Experimental results and discussion

One custom made CCECG setup was applied to all three cases. The only difference was the placement of the textile electrodes as seen in Fig. 8. A minimum of three

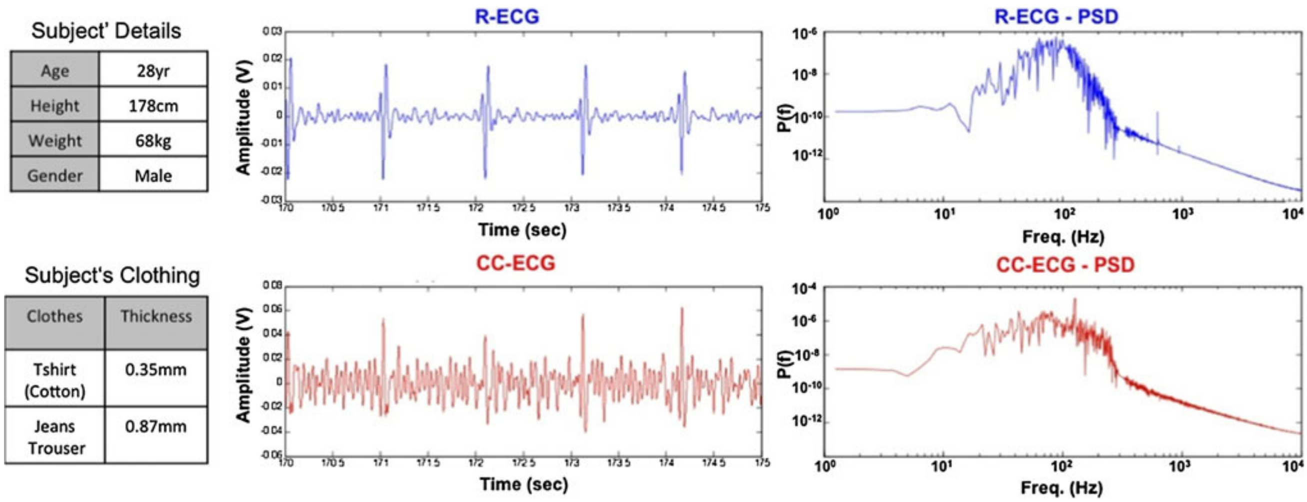


Fig. 10 ECG measurements on the hospital-bed for 1-min

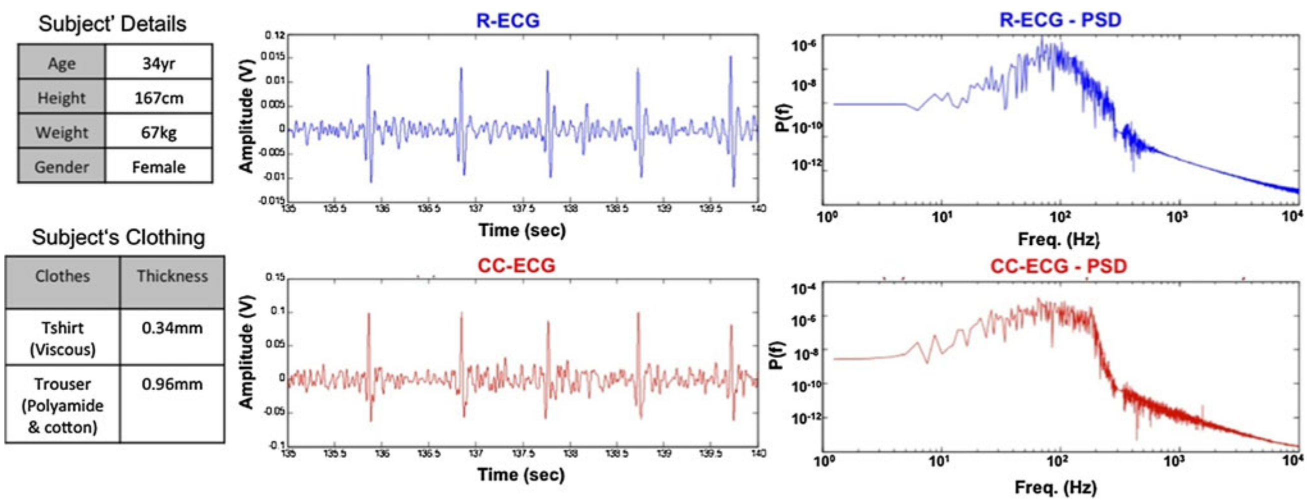


Fig. 11 ECG measurements in the wheel chair for 5 s

volunteers occupied each test setup for up to 30 min. The following sections explain the captured ECG data and also demonstrate the feasibility of Capacitive-Coupled Electrocardiography. A 2-lead contact, R-ECG was performed in parallel and used as a reference signal in order to compare with the CCECG signal. R-ECG and CCECG are plotted in all of the graphs with blue and red colors respectively.

5.1 Short-term ECG and spectrum analysis

The following section presents short-term data and simple analyses for each scenario in order to compare and characterize them. In each scenario, the subject's demographic and clothing details are listed.

5.1.1 Stretcher

Volunteers were asked to take the conventional supine position on the stretcher, as illustrated earlier. In this position electrodes were located posterior of the subject's heart. A detailed close-up of 5-s signals in Fig. 9 illustrates the time synchrony of capacitive and regular ECG signals. Even though the subject is wearing two layers of cloth, a T-shirt and a sweater, the amplitude of the CCECG signal lies in the same range as the R-ECG.

The power spectral density (PSD) plots illustrate similar frequency contents of both signals in the frequency bands up to 100 Hz. A steep decline in power above 200 Hz was found in CCECG signals with noise components at 300 Hz, which was not detected in the R-ECGR-ECG recordings.

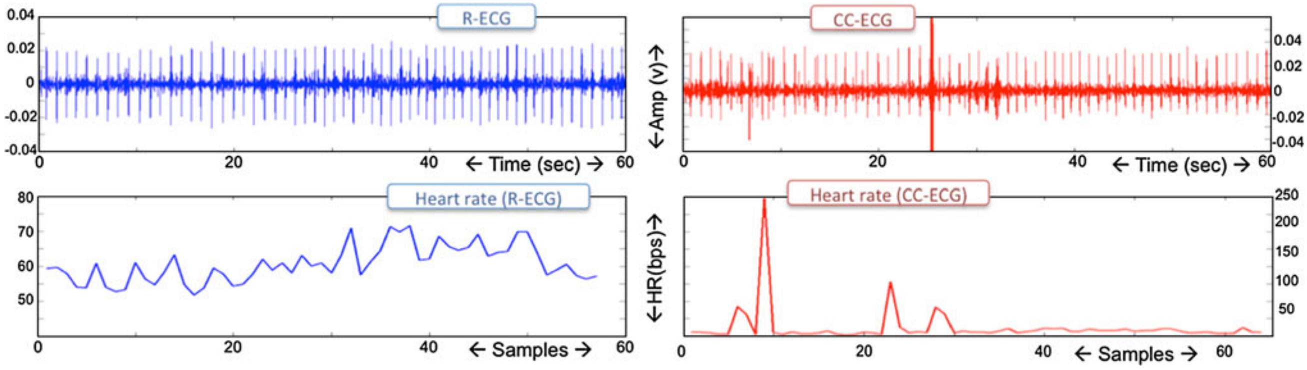


Fig. 12 ECG measurements on the stretcher for 1 min

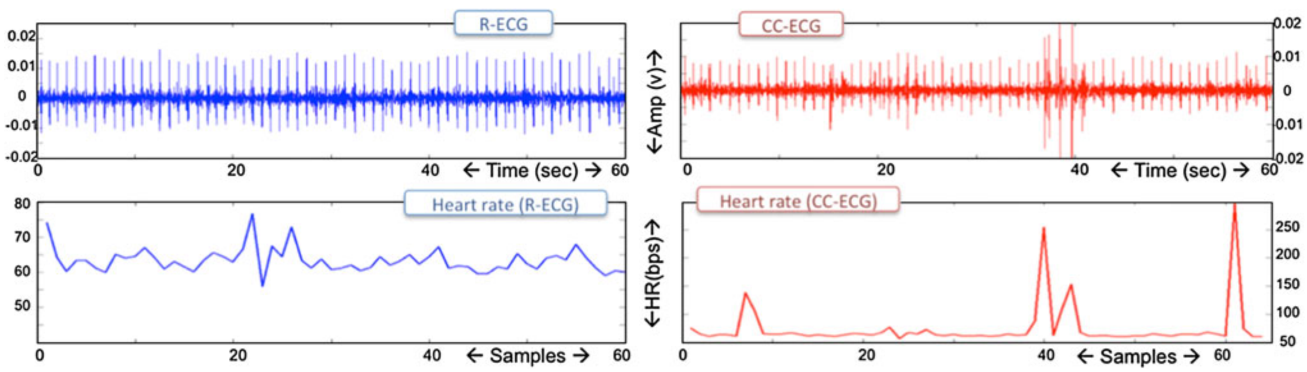


Fig. 13 ECG measurements on the wheelchair for 1 min

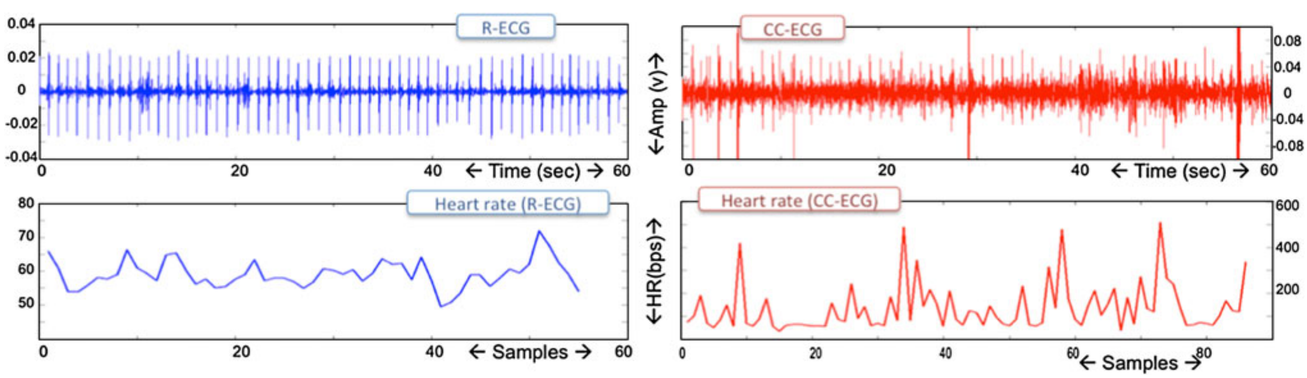


Fig. 14 ECG measurements on the hospital bed for 1 min

5.1.2 Hospital bed

Similar to the stretcher, volunteers lied down in a supine position emulating a patient on back in the clinical bed. CCECG electrodes were placed slightly separated and right behind the heart area of the subject.

Even though the soft bed-mattress is supposed to provide better contact between the textile electrodes and the body, the CCECG signal showed a higher noise level as the equivalent signal on the stretcher (Fig. 10). We hypothesized that the elastic mattress distorts the textile electrodes in an unforeseen way and hence makes the system more prone to environmental noise.

5.1.3 Wheelchair

In this scenario, subjects were asked to sit comfortably in the wheelchair and as relaxed as possible while holding the

Table 1 Comparison of QRS detections with CCECG raw signal and CCECG after considering refractory period (RP)

Setup	R-ECG	CC-ECG (raw)		CC-ECG (RP of 0.3 s)	
	QRS Count	QRS count	Accuracy (%)	QRS count	Accuracy (%)
Stretcher	57	61	92.98	59	96.49
Hospital-bed	55	86	61.81	71	70.9
Wheelchair	60	64	93.33	62	96.67

textile contacts for conventional ECG measurement. Figure 11 illustrates the results and shows similar results as found in the stretcher setup. The QRS complex detected in CCECG co-exists at the same time point of the QRS complex in R-ECG. Again, high frequency noise (>300 Hz) with low power is present in CCECG signal, but not in the R-ECG record.

5.2 QRS detection

This section presents 1 min ECG recordings of each clinical setup. QRS complexes from each recording have been detected with time of occurrences leading to the heart-rate, RR-interval timings and heart-rate sequences.

Figure 12 visualizes results of the stretcher setup and displays a 1 min recording of both R-ECG and CCECG. The R-ECG plot displays no obvious artifacts, while the CCECG plot shows the presence of large spikes and unwanted small spikes between regular RR intervals. Extremely high heart-rate values, as detected by the CCECG (e.g., at 8 and 23 s in the bottom right part of Fig. 12), illustrate the false positive QRS counts. Results are similar with few artifacts in case of the wheelchair scenario (Fig. 13).

The situation is a lot worse in the clinical bed scenario (Fig. 14), clearly requesting measurements to improve the QRS-complex detection in the CCECG recordings. The otherwise reliable post-processing of ECG signals suffers from severe artifacts, when performed with capacitively coupled electrodes. In any ways capacitive textile electrodes integrated on the hospital bed have been shown to potentially monitor patient's movement in order to reduce effect of pressure ulcers, common among long-term hospitalized patients (Yip et al. 2009).

The situation can be improved as shown in Table 1, when a refractory period of 0.3 s after the last QRS-complex is taken into account. The algorithm prohibits the false detection of the heartbeat within the refractory period. In each measurement setup, there was an improvement in accuracy, especially with the hospital-bed CCECG signal, which had low signal-to-noise ratio. Clearly this defeats the purpose of, for example, detecting extra-systoles in an ECG, but demonstrates the need for further investigating advanced signal processing in context with the CCECG.

6 Conclusions

In today's ageing society, the demand for unobtrusive health monitoring is growing, not the least due to a growing number of persons living in nursing homes or serviced apartments. Together with the presumed advantages of continuous ECG monitoring in the clinical environment, a need for contactless, non-obtrusive ECG recordings arises.

With this study, we showed the feasibility of contactless ECG measurements with a prototype capacitive-coupled ECG system. Our system uses conductive textiles, actively shielded by a driven seat circuit. Flexible textile electrodes were placed in three very common clinical settings. Here, the prototype was able to successfully record ECG signals from volunteers without the need to glue conductive electrodes on the subject's skin. Signals displayed in all three settings a good visual quality and thus later are used for further standard ECG analysis like QRS complex detection. The data presented strengthen the case for noise-reducing guarding layers to be used in CCECG. Our experiments demonstrated that the CCECG signals are prone to interference and movement artifacts, due to their delicate recording physics. This will become an obstacle in detecting QRS complexes from the signals later. In order to utilize contactless ECG measurements in locations of nursing, further steps to improve stability, signal quality and long-term acceptance have to be taken.

Even though the primary objective of our experiments was not to compare the subject's clothing and its effect on quality of ECG detection, we observed that ECG detection using the CCECG was hardly affected by the subject's clothing. In other words, thickness and material of the clothes worn by the subject didn't change quality of detected ECG signals. At one instance, the clothes not unexpectedly became a point of hindrance to the signal quality, when a subject was wearing a fur coat exceeding the minimal distance between skin and electrode. Based on these observations, we conclude that ECG detection quality of our system remains unaffected, when a subject is dressed for room temperature. Subsequently, as the layers of the garment increase, ECG detection quality deteriorates. However a rigorous experiment on the influence clothing has on the CCECG acquisition is beyond the scope of this and subject to a follow up study.

All in all, we interpreted our results as sufficiently promising to envision the potential future care facility featuring a personal, non-obtrusive and yet permanent health monitoring system providing reassuring baseline vital signs and fast response times against serious health problems.

Acknowledgments The authors would like to thank the BMBF (German Ministry for Education and Research) for funding the work of the INSITEX project. We are also thankful to TITV-Greiz, Greiz, Germany for providing us the textile-based contact electrodes. Our appreciations extend to Stephen Oung and Shivayogi Hiremath for their time and efforts on reviewing our manuscript.

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