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A Wireless Ballistocardiographic Chair

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Abstract— **This paper presents a wireless ballistocardiographic chair developed for the Proactive Health Monitoring project in the Institute of Signal Processing. EMFi sensors are used for BCG measurement and IEEE 802.15.4 RF link for radio communication between the chair and a PC. The chair measures two BCG signals from the seat and the backrest and a rough ECG signal from the armrests of the chair. The R-spike of the ECG signal can be used as a synchronisation point to extract individual BCG cardiac cycles. Also, two developed methods for extracting BCG cycles without using a reference ECG signal are presented and compared.**

I. INTRODUCTION

Ballistocardiogram [1] is a non-invasive technique for the assessment of the cardiac function. BCG consists mainly of heart movement and the movement of blood in aorta, arteries, and periphery. In our previous research, we have developed a normal-looking chair fitted with sensitive sensors for BCG measurement connected via cables to a commercial medical data-acquisition system [2], [3], [4], and signal processing algorithms for BCG analysis [5], [6], [7], [8]. In this paper, we present a wireless version of the measurement chair, and the improved version of the so-called blind BCG segmentation algorithm first introduced in [4].

In the developed wireless EMFi-BCG measurement chair, we used two EMFi-film[1] [9], [10] sensors to measure BCG from body movements in a sitting position. One EMFi sensor is installed under the upholstery on the seat panel and second one on the backrest. Only the seat sensor recordings are used in this paper. Copper electrodes are fitted to the armrests and are used to measure a coarse ECG signal from the wrists/hands of the person sitting on the chair. The hardware used for measurements is presented in this paper. Sample measurements and a small evaluation of the performance are presented. More information about the use of EMFi-sensor for BCG measurement can be found from [2], [4].

Usually electrocardiogram (ECG) and its R-spike are used as a reference to extract BCG cycles from the recorded signal. We present an improved blind segmentation method to extract BCG cycles per cardiac period without using the ECG signal as a reference. This method was developed using patient data recorded with a earlier wired prototype of the chair [2].

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[1]EMFi is a registered trademark of Emfit Ltd.

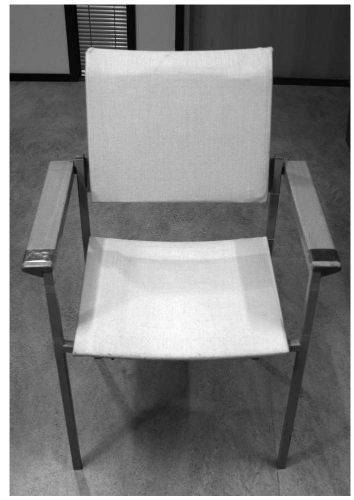

Fig. 1. The wireless EMFi-BCG measurement chair. The electronics are hidden under the chair. EMFi sensor films are fitted under the upholstery on the seat and the back panels of the chair. Small copper strips have been fitted to the armrests to measure the ECG.

II. MEASUREMENT ELECTRONICS

The measurement electronics are partly based on system developed earlier for a wired measurement chair prototype [2], [4]. They were improved and enhanced to make the amplifier for the wireless system. A Ballistocardiogram, or an EMFi-BCG signal, is measured from the seat and the backrest of the chair. A charge amplifier is needed to amplify the signal from the EMFi sensors. The charge amplifier is the same as used for the wired version. This is followed by a Sallen-Key filter circuit, which implements a second degree Butterworth low-pass filter with a cut-off frequency of 28 Hz. The BCG signal contains information between 0 to 40 Hz [11], and based on our research most of the information is between 1.5 to 20 Hz, which justifies using a low-pass filter instead of 50 Hz notch filter and a 125 Hz low-pass (for anti-aliasing).

Small copper strips (contact area of 4x6 cm) are installed on the ends of the armrests (see Fig. 1), and a coarse ECG signal can be measured from them using a sensitive ECG amplifier. The ECG amplifier is based on a circuit presented in [12], and its output is low-pass filtered using a simple first degree 36 Hz RC-low-pass filter. The amplifier and A/D board has also an amplifier for one Flexiforce[2] sensor, which would be used for measuring the weight of the person sitting on the chair. Currently this function is not used.

All analogue signals are connected to a 8-channel 16-

[2]Flexiforce is a registered trademark of Tekscan, Inc.

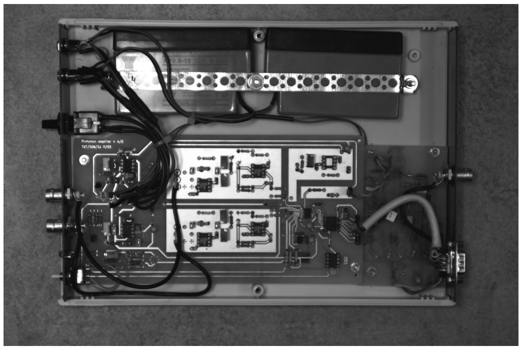

Fig. 2. The measurement electronics of the EMFi BCG chair. Two sealed lead batteries are used to power the amplifier, A/D converter and opto-isolator. The separated RF-board is attached to the serial type A/D converter via isolated serial interface.

bit serial A/D converter, the Texas Instruments ADS8345. Analog Devices ADR291 is used as a 2.5 V voltage reference. The SPI-interface controlling the A/D is isolated using ADUM1401 quad-channel digital isolator. The isolated signals are connected via cable to the R/F board. In addition, all amplifiers (except the ECG amplifier) and active filters were implemented using Analog Devices' AD820 opamps, and Maxim's MAX663 and 664 linear regulators were used for power supplies.

The amplifier is powered by two 12 V 0.8 Ah sealed lead-acid type batteries. The power consumption in idle-state (when no-one is sitting on the chair) is about 3.5 mA for positive supply battery and about 2.8 mA for negative supply. The 0.7 mA difference is mainly caused by the second power regulator circuitry connected to the positive supply battery, which powers the serial A/D converter and the isolator circuit. Also, the ECG amplifier uses only the positive supply, and some of the pull-up circuits drain the positive supply. The power consumption could be reduced for both batteries by using larger pull-up and pull-down resistors in some parts of the circuitry. The active power consumption (when a person is sitting on a chair) is about one to three mA more per battery. The whole measurement electronics unit with batteries is fitted in a 28x20x4 cm box installed under the chair. A picture of the box opened can be seen in Fig. 2.

III. RF-LINK

The RF-link has been implemented using ChipCon CC2420 chip designed for IEEE 802.15.4 / ZigBee applications. We used the CC2420DB development boards fitted with Atmel Mega128L microcontroller as the hardware platforms. A software SPI-interface was implemented using free I/O pins on the boards with additional enable-signal to control the A/D controller via a cable interface. The hardware SPI interface of the microcontroller is used to communicate with the CC2420. Data-communication is based on a partial implementation of the IEEE 802.15.4 MAC link. With 250 Hz sampling rate, the link is able to transmit four 16 bit values without problems. A full MAC version of the data-communication is under development, using ChipCon's IEEE

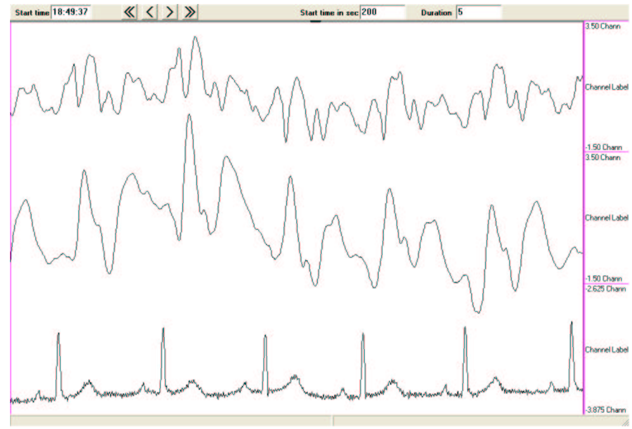

Fig. 3. Raw unfiltered seat and backrest BCG signals and armrest ECG displayed with EDFView.

802.15.4 MAC software. A partial implementation of the ZigBee stack working on top of the MAC is work in progress.

Another CC2420DB board is connected to a PC via the RS-232 interface. The PC has a real-time monitoring software implemented with Visual C++ running on Windows, which reads the data from the selected serial port and plots five seconds of data from four channels on-screen. The software is also able to save the data to files using the EDF file format. The files can then be viewed using the several existing EDF tools. Fig. 3 shows raw recorded BCG (2) and ECG signals on EDFView screen.

IV. BLIND SEGMENTATION OF BCG DATA

A recorded BCG signal consists of components attributable to cardiac activity, respiration, and body movements. To have a pure BCG signal and to remove additional components including background noise as well as respiration, we used a band pass filter with a passing band from 2 to 20 Hz. Body movements during recording corrupt some BCG cycles. These parts of the recorded BCG signal are useless and must be removed by using amplitude thresholds. This simple method is useful because body movements usually cause bigger signal changes compared to normal cardiac activities.

To extract BCG cycles from the recorded BCG signal without using any other synchronization signal, such as ECG, we have previously used an absolute value of the respiration signal (obtained by filtering the BCG signal) as presented in [4]. However, the problem with using respiration signal is the loss of some of the BCG cycles because of the time period of the respiration signal. To overcome this problem, we have extracted a coarse BCG signal using a narrow band pass filter with 1 and 2 Hz corner frequencies. In the same way as with the respiration signal, we used absolute values of this coarse BCG signal and its peaks within a lower and upper amplitude thresholds as synchronization points, eliminating peaks out of the defined range. Based on our experience, peaks out of this range are not related to BCG cycles, being background noises or motion artifacts. Uniform windows with a length of 250 samples (1.2 seconds) were used to extract individual BCG cycles. The synchronization points computed from the

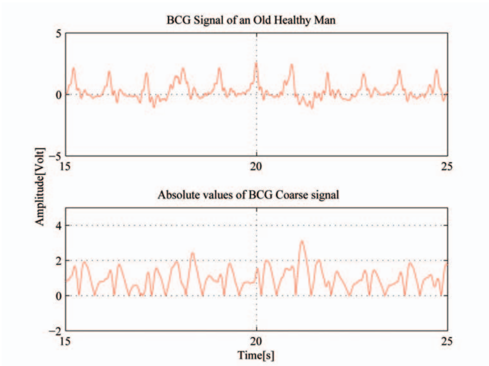

Fig. 4. Typical BCG and absolute values of the coarse BCG signal.

coarse BCG signal are used to find the central points of these windows and 125 samples before and 124 samples after these central points are taken to create BCG cycles with the same lengths. The cycle lengths are not adaptive to heart rate (HR), as we tried to extract the BCG cycles without the use of ECG (which could have been easily used to estimate the HR quite accurately), and the estimation of HR using only BCG was not accurate enough. Fig. 4 shows a sample BCG signal and calculated coarse BCG signal.

V. RESULTS

The EMFi sensor amplifier is designed to amplify a charge change signal. Measurement with voltage or current input signal produces a frequency response where 20 dB/dec falling response equals flat response. We used a signal generator with a series 10 MΩ resistor and measured the response using 10 and 50 mV sinusoidal input signals, and used Matlab to post-process the recordings to obtain true frequency response for charge change input. Because the amplifier has a huge gain for low-frequency signals and the armrest electrodes seem to pick up 50 Hz noise which interferes with weak signals at higher frequencies, the measurement of the frequency response with one fixed input amplitude would be a trade-off between low (0 to 2 Hz) and high (over 30 Hz) frequency accuracy. We used 10 mV input to measure the response from 0.1 to 40 Hz and 50 mV to measure the response from 7 to 125 Hz. We then scaled and combined the data between 0-15 Hz from the first and 16-120 Hz from the second recording to obtain the response depicted in Fig. 5. If compared to the frequency response of the previous battery-powered charge amplifier used in the wired implementation [4], which was relatively flat for the most interesting 2 to 20 Hz area, we can see that the new system does amplify the higher BCG components a bit more than the old one.

The linearity of the amplifier was also measured. The results (Fig. 6) show that the amplifier is linear with 3, 10 and 20 Hz test signals.

As can be seen from Fig. 7, the R component of the ECG can be clearly seen in the ECG signal recorded with the armrest electrodes and filtered with 50 Hz software notch filter. The raw unfiltered ECG can be seen in the EDFView screenshot in Fig. 3.

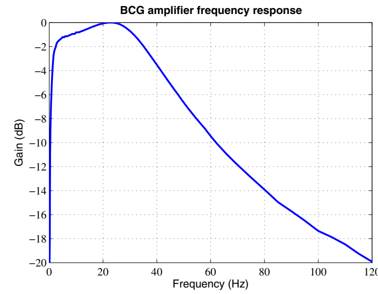

Fig. 5. Frequency response of the BCG amplifier.

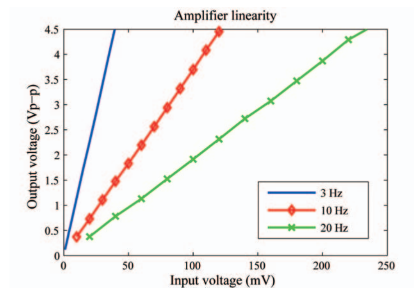

Fig. 6. Linearity of the BCG amplifier.

Table I shows the difference between numbers of extracted cycles using the two different blind segmentation method discussed in Section IV. Cycles were extracted from five minute BCG recordings of three subjects representing different patient groups. As can be seen, the new coarse signal based method is able to extract more cycles than respiration signal based. However, as the average heart rate of the subjects was around 60-75 bpm in all the recordings (about 300-375 beats per 5 minute recording), it can be seen that the new algorithm detects about three synchronization points per every two cardiac cycles. To verify this, we performed a one minute beat-to-beat comparison against recorded reference ECG. The results can be seen in Table II. For a typical young healthy man the method extracted 110 BCG cycles over one minute. The correct number of cardiac cycles was 75 of which the method extracted 74 cycles correctly. The other 35 cycles were repeated, redundant, and the same as previously extracted BCG cycles. We found also one undetected cycle, which was not detected because of a large motion artifact that destroyed a BCG cycle and caused the method to ignored it. Fig. 8 shows six extracted BCG cycles of an old healthy man, extracted using the coarse BCG method.

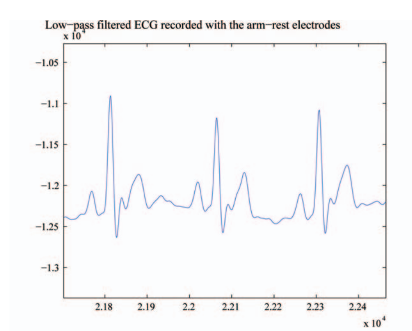

Fig. 7. Sample ECG signal recorded from the armrest electrodes.

TABLE I

RESULTS OF BLIND SEGMENTATION OF BCG DATA FOR THREE TYPICAL SUBJECTS (FIVE MINUTE BCG RECORDINGS)

Subject	N.R.S.	N.C.S.
Young healthy man	215	582
Old healthy man	271	571
Old man with past cardiac infarction	259	515

N.R.S. = Number of extracted cycles using respiration signal.
N.C.S. = Number of extracted cycles using coarse BCG signal.

TABLE II

ONE MINUTE BEAT-BY-BEAT COMPARISON OF CYCLES EXTRACTED USING COARSE BCG SIGNAL AND HEART CYCLE DETERMINED FROM REFERENCE ECG RECORDING

Subject	ECG cycles	Cardiac cycles extracted from BCG			
		Total	Correctly	Missed	Redundant
YHM	75	110	74 (99%)	1 (1%)	35
OHM	67	130	67 (100%)	0 (0%)	63
OMI	61	142	61 (100%)	0 (0%)	81

YHM = Young healthy man, OHM = Old healthy man,
OMI = Old man with past cardiac infarction.

VI. CONCLUSIONS AND FUTURE WORKS

A. Conclusions

In this paper, we presented a wireless ballistocardiographic chair, designed to look like a normal office chair. It can be used to measure EMFi-BCG signal from a person sitting on the chair without connecting any cables to the patient, and even without the patient knowing that he/she is being measured. The chair has electrodes fitted to the armrests to measure a coarse ECG signal for R-spike position detection, but in this paper we have shown, that it is also possible to extract most of the BCG cycles without the ECG reference and thus without the ECG electrodes.

The obtained results show that the performance of the segmentation method based on coarse BCG signal is high, but it creates a lot of redundant BCG cycles, e.g., many of the cycles are extracted twice. The method could still be used in pattern recognition or similar applications which tolerate redundant cycles.

B. Future Works

The new amplifier amplifies the higher (around 20 Hz) BCG components slightly more than the lower components.

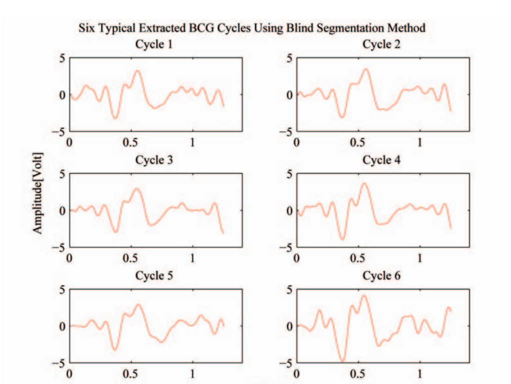

Fig. 8. Typical six BCG cycles of an old healthy man, extracted using the coarse BCG signal.

As the charge amplifier is the same as in the old system, this behavior is most likely caused by the Sallen-Key low-pass filter added to the system. The system could be improved to obtain even more flatter frequency response in the 2 to 20 Hz band. The ECG signal is very weak, and needs lots of software amplification. The gain of the ECG amplifier should be increased.

The current blind BCG cycle extraction method finds too many synchronization points and has to be improved. A synchronization point is only used as starting point to find the central point of one BCG cycle, so the false synchronization points produce copies of some cycles, but not actually new false cycles.

VII. ACKNOWLEDGMENTS

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