A portable system for acquiring and removing motion artefact from ECG signals

Adrian Griffiths, Aruneema Das, Bosco Fernandes and Patrick Gaydecki

School of Electrical and Electronic Engineering, University of Manchester, PO Box 88, Manchester M60 1QD, United Kingdom

E-mail: patrick.gaydecki@manchester.ac.uk

Abstract. A novel electrocardiograph (ECG) signal acquisition and display system is under development. It is designed for patients ranging from the elderly to athletes. The signals are obtained from electrodes integrated into a vest, amplified, digitally processed and transmitted via Bluetooth to a PC with a Labview ® interface. Digital signal processing is performed to remove movement artefact and electromyographic (EMG) noise, which severely distorts signal morphology and complicates clinical diagnosis. Independent component analysis (ICA) is also used to improve the signal quality. The complete system will integrate the electronics into a single module which will be embedded in the vest.

1. Introduction

Heart patients sometimes experience events such as arrhythmias and inadequate blood flow to the heart muscle. These occur unpredictably and last for very short periods of time. In order to study such events, it becomes necessary to continuously monitor the patient’s ECG activity over a period of time. Clinicians usually use a Holter monitor to collect data from patients over a period of 24 hours. The system consists of a battery powered unit to which leads from electrodes attached to the body are connected. The unit records ECG signals on a memory card, whilst the patient makes notes of any noticeable symptoms during the testing period. After the monitoring period, the sets of data are analysed.

In this project, a system is being developed which will enable patients to monitor their heart functions with an intelligent garment. The garment is a specially designed, closely fitting vest, which incorporates four electrodes. They are connected to a module which contains amplification, digital signal processing hardware and wireless communication sub modules. The ultimate aim of this work is to integrate this module into the vest. It is intended that the data will be able to be transmitted wirelessly to a PC and sent via the internet for remote analysis.

2. Theoretical considerations

The heart has four chambers, comprising the atria at the top half and the ventricles at the bottom half. The right atrium receives deoxygenated blood from the body and transfers it into the right ventricle, from where it is pumped into the lungs, to release carbon dioxide and be replenished with oxygen. The oxygenated blood feeds into the left atrium and is transferred into the left ventricle from where it is forced into the aorta and the rest of the body.
The heart muscle is made to contract and therefore pump blood around the body by means of an autonomous electrical signal that is generated by a bundle of specialized muscle fibres called the sinoatrial (SA) node. This node is located in the upper region of the right atrium. Within the node, changes in ionic concentration across the cell membranes cause a propagation of electrical activity to occur. The potentials build up to form an activation wavefront which manifests itself as a P wave, the first segment of an ECG signal, when it encounters the wall of the atrial muscle. This impulse proceeds through the atrio-ventricular (AV) node and a bundle of nerves around the ventricular region called the His-Purkinje system [1]. There is a short period of isoelectric activity following the P wave after which the muscles of the ventricles are excited, causing a large deflection of the heart and a spiked waveform in three segments. The initial downward deflection is the Q wave, the large upward deflection is called the R wave and the final downward deflection is called the S wave. This section is usually referred to as the QRS complex. Following this complex is another relatively short isoelectric segment followed by the T wave, when re-polarization occurs and the ventricles return to their resting state. There is also another wave which may occur in some individuals as part of the re-polarization process, called the U wave. The left hand side of Figure 1 shows the components of one full ECG cycle.

![Diagram showing the PQRST complex of an ECG signal and the electrode positions on a human torso.](image)

**Figure 1.** Diagram showing the PQRST complex of an ECG signal and the electrode positions on a human torso.

It is possible to non-invasively record the electrical activity of the heart by placing electrodes at specific points on the body. This was first proven by Einthoven around 1900 [1]. Three main methods have now been established for recording ECG signals from patients. One uses a 12 lead system where electrodes are placed around the heart. The second is known as a 3 lead system, where three electrodes are placed on the torso in the following orientation; one near the right arm (RA), just below the right clavicle, one near the left arm (LA), one just below the left clavicle and one above the left leg (LL), above the hip. There is also another reference electrode placed above the right hip, called the indifferent electrode. This arrangement is known as the Einthoven triangle and is shown on the right side of Figure 1. The measurement across the RA and LA is referred to as Lead I. The RA and LL measurement position is referred to as Lead II and the LA and LL measurement position is referred to as Lead III. The method used for monitoring long term or ambulatory patients is by either one or two lead system. This is usually sufficient for monitoring life threatening disruptions to the ECG. In this work, the system under development currently monitors only one lead at a time.

Electrode design commonly consists of a silver conductor which connects to the body. An electrolyte of Silver chloride forms the electrical interface between the metal electrode and the skin. The central gel area of the electrode is coated with adhesive to enable the electrode to be fixed to the skin. In this work, the electrodes are part of the garment.
3. System design

The schematic diagram in Figure 2 shows the complete portable wireless ECG system along with the vest and a laptop for displaying the received ECG data, in real time. The current system design allows one channel of ECG data to be obtained from a subject wearing the vest. The integrated electrodes are sited to represent the Einhoven triangle and thus only the simplest bi-polar electrocardiographic leads are used [2]. These standard leads consist of electrode pairs, which detect the electrical potential between the two points across the subject’s heart.

![Block diagram of portable real time DSP system for wireless ECG.](image)

The electrical potential from cardiac activity is low, in the millivolt range, and the heart rate is normally less than 200 beats per minute (BMP). To obtain a useful signal for acquisition and processing the first stage of the system uses a precision analogue instrumentation amplifier. The main amplifier considerations in this application are the need for a high differential gain, low common mode gain, high common mode rejection ratio (CMRR) and high input impedance. The bandwidth is not critical, since the required frequency range lies between 0.05Hz – 100Hz [2]; sampling above 200 Hz is therefore sufficient to obtain the full morphology of the ECG signal.

The characteristics of the amplifier and the associated circuitry provide a controllable gain, in the range between \( \times 1000 \) and \( \times 5000 \). This ensures the signal has a voltage range large enough to be digitised successfully and utilizes the full 10 bit resolution of the analogue to digital converter (ADC). The circuitry also allows the output signal to be offset to ensure the signal swings between the supply rails. This is achieved manually at present, but could be designed to be automatic in the future. The differential input impedance is high, in the order of \( 10^9 \Omega \), with a CMRR in excess of 70dBs.

The amplified analogue ECG signal will also contain contaminated noise from a variety of sources. These include EMG muscle movement, EMI environmental noise and movement artefact created from the electrode surface moving against the subject. To eliminate this interference the signal is passed to a commercially available real time digital signal processing unit that was designed within the group.
called ‘Signal Wizard ™’. This unit is used to design very sharp digital finite impulse response (FIR) filters which are run in real time. The system can also be used to implement infinite impulse response (IIR) and adaptive filters. These techniques are currently being investigated to remove motion artefact.

Once filtered, the data are passed to a microcontroller that communicates with a Bluetooth wireless transceiver. The data are transferred serially at bit rates ranging from 9600 (default) to 57600, which are selectable via switch settings on the circuit board. The serial data rate also dictates the acquisition sampling rate of the data, which for 9600 baud and two bytes of data is 480 samples per second. The Bluetooth transceiver is capable of full duplex mode, but is only used for transmission in this application; it is also a class 1 Bluetooth radio, allowing transmission distances up to 100 metres. The unit has the ability to multiplex a second channel; although this would halve the sampling rate, this could be compensated by increasing the baud rate. The second channel is currently unused and has been reserved for use with other sensors in the future, such as those for respiration monitoring.

Reception of the ECG data is achieved using a PC with Bluetooth capability. The data acquisition software written in Labview ™ synchronises the serial reception and displays the data in a graphical form on the screen, in real time. The data can also be logged to file for offline analysis. A screenshot of the interface is shown in Figure 3.

Since the system is attached to a human subject, all the electronics are battery powered. Current limiting resistors are connected in series and low forward voltage drop Germanium diodes are connected at the input stage of the amplifier to provide isolation from the patient. This system does not feed any current to the patient.

4. Experimentation and results
The system has been tested in the laboratory and has been used to obtain data from a vest with the subject both stationary and with movement of the arms. Figure 4 shows the equipment and Figure 5 shows the raw data that were obtained. In Figure 6, the data have been passed through a band pass FIR filter with a lower cut-off of 0.6 Hz and an upper limit of 40 Hz.
Figure 4. Photograph of the test equipment showing the vest, acquisition electronics, battery pack, DSP filter and notebook PC

Figure 5. Raw data collected by the wireless system from a stationary subject. The trace contains low frequency drift from respiration

Figure 6. Data processed by band pass filtering to remove low frequency drift due to respiration

5. Post-processing with ICA
Morphology of the ECG signal is very important for the purpose of clinical diagnostics associated with it. Motion artefacts and noise arising from respiration, muscle movements, electrode movements, electrical noise from surrounding and the likes detrimentally affect the morphology of the measured ECG signal. Many methods have been used for the removal of noise and the artefacts from the ECG signal. Recently many works were carried out in the field of blind source separation (BSS) using a new technique called independent component analysis (ICA). It is a source separation algorithm and used widely in biomedical signal processing applications. ICA is a statistical technique for decomposing a mixed dataset into independent components; it is predicated on the assumption that the components are statistically independent and non-Gaussian [3-5]. This work aims to use ICA coupled with windowing to reduce noise and artefacts from ECG signal to a considerable extent rendering it suitable for diagnostics purpose.

To describe the ICA algorithm each of the original independent signals will be referred as $s_i$ and each of the linearly combined mixed signals as $x_i$ where $\mathbf{x}$ is a column vector of $n$ measured signals. Each measured signal can be expressed as a linear combination of the original independent signals:
We can express the entire system of $n$ measured signals as:

$$x = As$$  \hspace{1cm} (2)$$

where each row of $x$ is a set of readings for each signal $x_i$, each row of $s$ is an original signal $s_i$, and $A$ is an $n \times n$ mixing matrix that generates $x$ from $s$.

The goal of ICA is, given $x$, find $s$ and $A$. ICA transforms the observed data $x$, using a linear static transformation $s = Wx$, into maximally independent components $s$. This is done by calculating the $w$ vectors and setting up a cost function which either maximizes the non-gaussianity of the calculated $s = (w^T * x)$ or minimizes the mutual information. The original sources $s$ can be recovered by multiplying the observed signals $x$ with the inverse of the mixing matrix $W = A^{-1}$, also known as the unmixing matrix. General parameters for measuring the non-gaussianity or statistical independence of the component sources include kurtosis and approximations of negative entropy (sometimes termed negentropy). The most popular algorithm for ICA is FastICA and it has been used in this work [6, 7].

In the ICA tests, the 3 channel ECG data was 51 seconds in duration and sampled at a rate of 2 kHz, with each channel containing 3 distinct segments. It was obtained from a separate system capable of recording more than one lead configuration simultaneously. The 1st segment of the data represents the subject at rest, the 2nd segment represents the subject jogging on the spot and the 3rd segment is of the subject at rest after the jogging. The data were filtered during collection to remove the base level shifts. The 3 channel ECG data is shown in Figure 7. It was further pre-processed by whitening before being fed to the FastICA algorithm. Whitening ensured that all dimensions were treated equally before the algorithm was executed. The signal processing was performed in MATLAB which was then incorporated into a Labview module.

It can be seen from Figure 7 that channel II and channel III contains a large amount of noise in the jogging segment and the ECG morphology is lost. The task of ICA was to reduce the noise and produce an estimate of the original ECG signal as accurate as possible.

![Original noisy ECG signals](image)

**Figure 7:** The original noisy ECG signal.

During the analysis, it was found that better results were achieved when the data were separated into segments, i.e. walking and jogging. A windowed ICA approach was adopted. When windowing was used to separate the noise free or less noisy segments from the noisier segments with FastICA
being applied to the noisier segment, it reduced the noise to a considerable amount, rendering the ECG signal fit for diagnostic purposes as shown in Figure 8. As noise and artefacts were present in the entire part of the windowed data segment, the un-mixing matrix could be rightly formulated to produce the estimation of source components of that segment. Windowing of long, complex and noisy data is an essential pre-processing task prior to the application of the ICA algorithm to reproduce the original source components.

Figure 8: ICA output of the windowed 2nd segment of each data channel

6. Conclusion
The system developed thus far has enabled the team to obtain clinically acceptable ECG signals from a vest using a wireless system. Digital filtering and ICA based post-processing have enabled respiration drift and movement artefact to be significantly reduced. The tasks that remain are further refinement of the software and miniaturization of the electronics into a single module, so that it can be integrated into the vest. Upon successful completion of this phase, the system will be improved to incorporate other sensors for respiration and temperature.

Acknowledgment
The authors wish to express their thanks to the SmartLife Ltd. for financially supporting this work.

References
[1] Tompkins, WJ, Biomedical digital signal processing: C-language examples and laboratory experiments for the IBM PC, 24-54 (1993)
[2] Webster John G, Medical instrumentation : Application and Design (Wiley, NY) ISBN: 0-471 153680 (1998)
[3] Comon P, Independent component analysis—a new concept? Signal Processing, 36:287-314 (1994).
[4] Hyvärinen A, Survey on Independent Component Analysis. Neural Computing Surveys 2:94 128 (1999).
[5] Hyvärinen A, Karhunen J and Oja E, Independent Component Analysis, John Wiley & Sons (2001).
[6] Hyvärinen A and Oja E, A Fast Fixed-Point Algorithm for Independent Component Analysis, Neural Computation, 9(7):1483-1492 (1997).
[7] Hyvärinen A and Oja E, Independent Component Analysis: Algorithms and Applications, Neural Networks, 13(4-5):411-430 (2000).