PORTABLE EQUIPMENTS FOR REMOTE MONITORING OF HEART ACTIVITY IN TELEMEDICINE

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Abstract

In this paper we describe the software development of a 12-lead ECG.

The device is designed to capture the 12-lead ECG and transmit it via Bluetooth to a standard personal computer. The personal computer can then be used to store, display or print the recorded ECG.

The software was developed using Visual Basic and is designed to run on any device supporting the .NET Framework.

Further work is required to refine the developed software to support enhanced visualization and storage of the recorded data.

1. Introduction

Bluetooth technology is intended to replace the cables connecting portable and/or fixed devices while maintaining high levels of security.

A fundamental strength of Bluetooth wireless technology is the ability to simultaneously handle data and voice transmissions which provides users with a variety of innovative solution such as hands-free headsets for voice calls, printing and fax capabilities, and synchronization for PCs and mobile phones, just to name a few. The range of Bluetooth technology is application specific.

Bluetooth technology operates in the unlicensed industrial, scientific and medical (ISM) band at 2.4 to 2.485 GHz, using a spread spectrum, frequency hopping, full-duplex signal at a nominal rate of 1600 hop/sec. The 2.4 GHz ISM band is available and unlicensed most countries.

Bluetooth technology's adaptive frequency hopping (AFH) capability was designed to reduce interference between wireless technologies sharing the 2.4 GHz spectrum. AFH works within the spectrum to take advantage of the available frequency. This is done by the technology detecting other devices in the spectrum and avoiding the frequencies they are using. This adaptive hopping among 79 frequencies at 1 MHz intervals gives a high degree of interference immunity and also allows for more efficient transmission within the spectrum. For users of Bluetooth technology this hopping provides greater performance even when other technologies are being used along with Bluetooth technology.

The EKG device detects and amplifies the tiny electrical changes on the skin that are caused when the heart muscle depolarizes during each heart beat.

Devices on the market that analyze ECGs, such as patient monitors, stress test systems, and holter analysis systems, do a good job of detecting beats and classifying arrhythmias.

This software implements the basic ECG analysis functions of beat detection and classification as C function. This release includes three version of beat detector. Two are general-purpose beat detection, where one represents a more efficient version with slightly different performance characteristics. The third beat detector is more computationally efficient, uses very little memory, and is embedded in a program that performs beat classification functions and may be used alone in applications that do not require beat classification.

2. Main text

Fig. 1-1 shows the basic operations of the beat detection algorithm. The beat detection algorithm can be broken down into two sections, the filters and the detection rules. This present release contains three versions of QRS detector. All three versions use the same filters and differ primarily in the details of the detection rules and code implementations.

The filters signal to generate a windowed (time limited) estimate of the energy in the QRS frequency band.

- Low pass filtering,

- High pass filtering,
- Taking the derivative,
- Taking the absolute value of the signal
- Averaging the absolute value an 80 ms window.

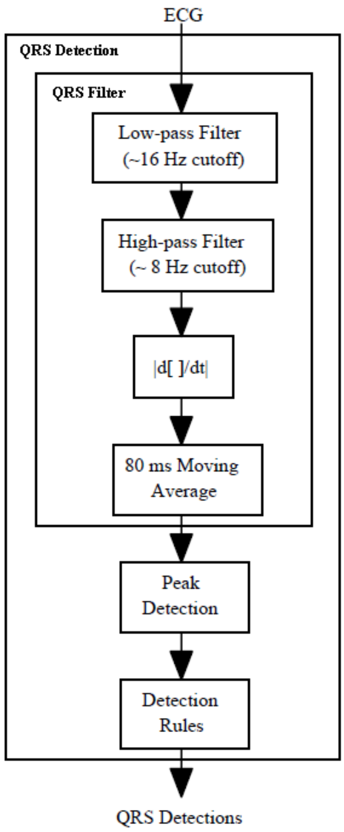

Figure 1-1. Beat detection operations

The final filter output produces what might be called a lump every time a QRS complex occurs. T-waves generally produce smaller lumps than QRS complexes. The high pass, low pass, and derivative combine to produce a bandpass filter with a pass band from 5 to 11 Hz, roughly the bandwidth that contains most of the energy in the QRS complex. The theory and implementation of these filters are detailed in [1], [2], and *Biomedical Digital Signal Processing* [3].

In [1] and [2], the filtered signal was squared rather than rectified. This operation caused the QRS detector to be somewhat gain sensitive. In this implementation I have used the absolute value, reducing the gain sensitivity and slightly improving the performance of the algorithm. The averaging windows was chosen to be roughly the width of a typical QRS complex. In the original algorithm this window was 150 ms wide to allow for the wide QRS complexes produced by Premature Ventricular Contractions (PVCs). Since then, it has been shown that a narrower window produces better results [4].

After the signal has been filtered, QRS detects peaks in the signal. Each time a peak is detected it is classified as either a QRS complex or noise, or it is saved for later classification. The algorithm uses the peak height, peak location and maximum derivative to classify peaks. The following is an outline of the basic detection rules for the algorithm.

1. Ignore all peaks that precede or follow larger peaks by less than 200 ms.
2. If a peak occurs, check to see whether the raw signal contained both positive and negative slopes. If not, the peak represents a baseline shift.
3. If the peak occurred within 360 ms of a previous detection check to see if the maximum derivative in the raw signal was at least half the maximum derivative of the previous detection. If not, the peak is assumed to be a T-wave.
4. If the peak is larger than the detection threshold call it a QRS complex, otherwise call it noise.
5. If no QRS has been detected within 1.5 R-to-R intervals, there was a peak that was larger than half the detection threshold, and the peak followed the preceding detection by at least 360 ms, classify that peak as a QRS complex.

The rules as outlined above are implemented in QRSDET and detailed in [1] and [2].

The detection threshold used in 4 and 5 above is calculated using estimates of the QRS peak and noise peak heights. Every time a peak is classified as a QRS complex, it is added to a buffer containing the eight most recent QRS peaks. Every time a peak occurs that is not classified as a QRS complex, it is added to a buffer containing the eight most recent non-QRS peaks (noise peaks). The detection threshold is set between the mean or median of the noise peak and QRS peak buffers according to the formula:

Detection_Threshold = Average_Noise_Peak + *TH**(Average_QRS_Peak Average_Noise_Peak)

where *TH* is the threshold coefficient. Similarly, the R-to-R interval estimate used in 5 is calculated as the median or mean of the last eight R-to-R intervals. Originally, I estimated average QRS peak values, noise peak values, and average R-to-R intervals using the median of the last eight values. The beat detector must begin with some initial threshold estimate. In order to make an initial estimate, I detect the maximum peaks in eight consecutive 1-second intervals. These eight peaks are used as are initial eight values in the QRS peak buffer, I set the initial eight noise peaks to 0, and I set the initial threshold accordingly. I initially set the eight most recent R-to-R intervals to 1 second.

3. Illustrations

The proposed ECG telemedicine system is divided in several functional blocks: a DAM, an Access Point and local or remote station supporting telemedicine software application Figure 3-0. Firstly, the DAM is the device in charge of the acquisition, digitalization and processing of patient's ECG signals. Once the data is processed, it can be stored in a memory card for posterior inspection, or it can be wirelessly transmitted via Bluetooth to an AP or via USB to a local monitoring station [8].

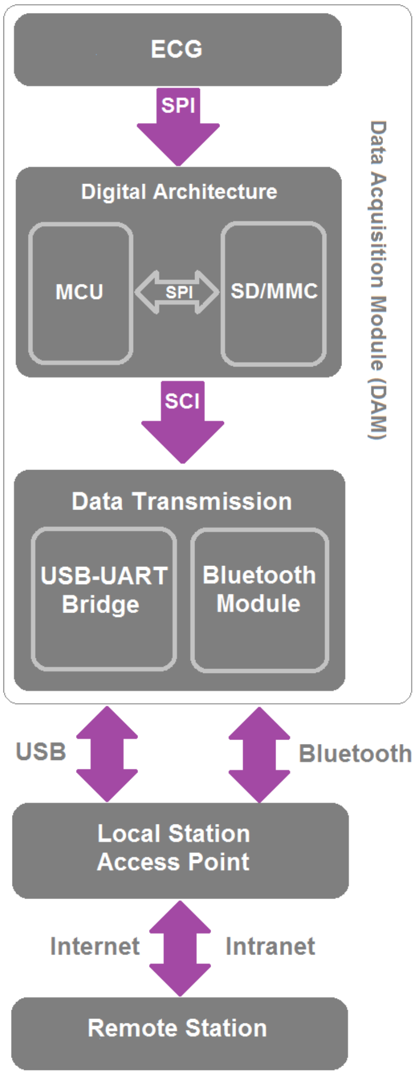

Figure 3-0

Secondly, the AP is a device located in a place near the patient which maintains connection between the DAM and the remote station via Internet/Intranet using TCP/IP. This connection could be continuous or event driven, t.e. when a risk situation is detected [9].

Finally, the local and remote stations run applications to visualize, analyze and store the information received from each patient.

UART interface is a standard 4 – wire interface with adjustable baud rates from 1200bps to 3Mbps [6].

3.1. Figure and table captions

Bluetooth supports multiple connections up to 4 slave units. There are two types of multiple connection modes: Multi-Drop Mode and Node Switching Mode.

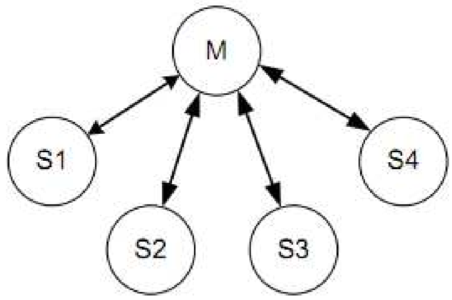

Figure 3-1. Multi-Drop Mode

In Multi-Drop Mode a master unit can connect to maximum 4 slave units at the same time and they transfer data bi-directional as in Figure 3-1.

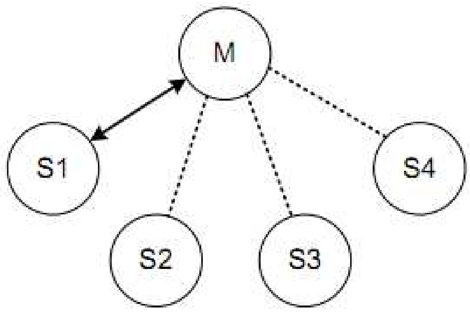

Figure 3-2. Node Switching Mode

In Node Switching Mode, the master unit maintains multiple connections with maximum 4 slave units but only one connection with one slave unit is active and data is transferred as shown in Figure 3-2.

Bluetooth compatibility is attained with an OEM module from BlueRadios, the BR-C30 Class 1[5]. This module is configured using AT commands and accepts or establishes connections with other devices using Serial Port Profile (SPP) conforming to Bluetooth V1.2. The default communication baud rate is 115.2 kbps.

Performance at different sample rates-all filter lengths and time related constants scale with sample rate changes so that the constants and filter lengths are equivalent to the constants and filter lengths in the 200/100 samples-per-second implementation. Ideally, implementations at other sample rates would perform the same as the 200/100 samples-per-second implementation, but round off approximations in constants and filter lengths result in slight differences in performance [7].

Table 3-1 lists the sensitivities and positive predictivities for beat detection and beat classification for three combinations of base rate and beat rate.

Base Rate	Beat Rate	QRS Sens.	QRS +Pred.	PVC Sens.	PVC +Pred.
200	100	0.9975	0.9981	0.9371	0.9664
250	125	0.9974	0.9981	0.9359	0.9597
300	150	0.9974	0.9979	0.9303	0.9665

Table 3-1. Beat Detector and Classifier Performance at Different Sample Rates

Table 3-2 lists the sensitivities and positive predictivities for beat detection alone on a wider range of sample rates [10]. Performance differences only seem significant when the base sample rate is dropped as low as 100 or 125 SPS [11].

Sam. Rate	QRS Sens.	QRS +Pred.	QRS Sens.	QRS +Pred.
100	0.996856	0.997905	0.995839	0.996423
125	0.997426	0.998257	0.99666	0.996788
150	0.997458	0.998016	0.997429	0.997652
175	0.997601	0.998093	0.997119	0.997268
200	0.997426	0.998071	0.997397	0.997588
225	0.997228	0.997994	0.997268	0.997769
250	0.997502	0.99806	0.997087	0.9973
300	0.99736	0.998016	0.99745	0.997897
325	0.997448	0.997874	0.997578	0.997684
360	0.997535	0.998038	0.997503	0.997865

Table 3-2. Beat Detector Performance at Different Sample Rates

4. Conclusion

An 12-lead ECG telemedicine device for non-clinical applications has been successfully developed and tested for two particular functions: ecg-holter and on-line transmission. Since, all memory cards tested exceed the usual sampling rate for holter mode, the TransFlash was selected because it is smaller than the others. Further study will be carried out to test transmission reliability in the presence of a variety of standard consumer electronics, e.g. cordless phones and WiFi devices operating in the same frequency band that may adversely affect data transmissions via Bluetooth at different distances.

Furthermore, data compression algorithms will be studied to optimize memory card usage by reducing records size and power consumption due to data writing. Finally, an agent-based Java application is being developed to provide analysis and detection of arrhythmia using several vital-sign signals.

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