AN INTERNET-BASED REAL-TIME DSP SYSTEM IN THE APPLICATION OF TELEMEDICINE

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Abstract

Telemedicine systems represent significant achievements in the provision of clinical medicine and health care service using telecommunication and information technologies for the purpose of remote monitoring. Almost all telemedicine systems require a network-enabled device, ranging from server machine to smart phone, which delivers the data as a transmission gateway.

The research in this thesis introduces the hardware and software design of a novel Internet-based real time DSP system in the application of telemedicine. Before this work, it was not previously achievable or economically feasible to develop a telemedicine system with a truly embedded measurement platform for real-time monitoring of clinical information on a global scale. The novelty of this design consists in embedding the Internet-based monitoring into the real-time signal processing system, as well as incorporating the merits of wireless communication and global distributed measurement.

To demonstrate this concept, a prototype of a truly embedded device incorporating either a browser-based application or a LabVIEW software application has been designed and developed, which is able to provide real-time biomedical signal acquisition, processing, wired/wireless transmission, visualisation, storage and retrieval via the Internet. The concept-to-prototype manipulates multiple biomedical signals from multi-sensors during studies and distributes them to the Internet. The prototype was evaluated on volunteers *in vivo* under ethical approval. The designed system was also tested under various physiological conditions and different Internet speeds. It manifests desirable performance regarding multi-functionality, ubiquitous accessibility, robustness, and adaptability. The full functionality of this innovative system successfully enables clinicians to remotely monitor a patient's physical condition in real-time globally. The experimental results obtained from the host are in close agreement with the expected performance of the designed system, which proffers evidence that this system represents a true innovation in the realm of telemedicine.

DECLARATION

That no portion of the work referred to in the thesis has been submitted in support of an application for another degree or qualification of this or any other university or other institute of learning

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My dream is to make my parents, my wife, my son and my country proud of me

To my parents, Zhu Shiwei and Li Minzhen;

to my wife, Liu Yu;

to my son, Zhu Chenyou;

for their endless love, encouragement and support.

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Glossary

8P8C	8 position 8 contact
ACK	acknowledgement character
ADC	analogue-to-digital converter
Ag/AgCl	silver/silver chloride
AHA	the American heart association
AV	atrioventricular
aVF	Lead augmented vector foot
aVL	Lead augmented vector left
aVR	Lead augmented vector right
BDM	background debug mode
BPM	beats per minute
CDMA	code division multiple access
CMRR	common mode rejection ratio
CRC	frame check sequence
DAQ	data acquisition
DHCP	dynamic host configuration protocol
DSL	digital subscriber line
DNS	domain name server
DSP	digital signal processing
ECG	electrocardiograph

EEG	electroencephalograph
EMG	electromyography
EMI	electromagnetic interference
EPHY	Ethernet physical transceiver
FEC	fast Ethernet controller
FFS	flash file system
FFT	fast Fourier transform
FHSS	frequency hopping spread spectrum
FIR	finite impulse response
FTP	File Transfer Protocol
GPRS	general packet radio service
GSM	global system for mobile communications
GSR	galvanic skin response
GUI	graphical user interface
HR	heart rate
HTTP	Hyper-Text Transport Protocol
ICMP	Internet control messaging protocol
ICT	information and communication technologies
IIR	infinite impulse response
IP	Internet Protocol
ISO	the international organisation for standardisation
LAN	local area network

LLC	logical link control
MAC	media access control
MII	medium-independent interface
OSI	open systems interconnection
PDA	personal digital assistant
PHY	physical layer
PPG	photoplethysmogram
QoS	quality of service
RF	radio frequency
RFI	radio frequency interference
RFID	radio frequency identification
RMS	root mean squared
SA	sinoatrial
SMS	short message service
SNR	signal-to-noise ratio
SoC	System-on-Chip
SpO_2	saturation of peripheral oxygen
SRAM	static random access memory
TCP	Transmission Control Protocol
TDMA	time division multiple access
UART	universal asynchronous receiver/transmitter
UDP	User Datagram Protocol

WAN	wide area network
WLAN	wireless local area network
WPAN	wireless personal area network

CHAPTER 1

Introduction

The subject of this research programme is concerned with design and establishment of an innovative and advanced telemedicine system. The main objective of this study focuses on the hardware and software design and development of an Internet-based real-time digital signal processing system in the application of telemedicine. The purpose of the hardware is to enable a truly portable embedded device. The purpose of the software is to allow ubiquitous accessibility by means of the Internet. The envisioned system is capable of performing acquisition, processing, wired/wireless transmission, display, storage and retrieval of clinical biomedical signals, which enables clinicians to monitor a patient's physical condition via the Internet, on a global scale. The designed system aims to reduce hospitalisation in the monitoring of patients, to yield a better distribution service anytime, anywhere and to improve quality of home healthcare.

Biomedical signal acquisition and processing is one of the fastest growing fields in the science and technology of digital signal processing (DSP) (Varshney, 2009, Bronzino, 2000, Gaydecki, 2004, Maheu et al., 2001, Goldberger. and Ng., 2010). Not all data from the real world is always instantly express valuable information in an accessible form. Processing, analysis and manipulation of signals are essential requirement in many engineering applications. Hence, DSP has a most vital impact in nearly every area where the information is manipulated in digital form. This is no overstatement – the modern world in which we live is in the digital information age. DSP is the engine of the technological and revolutionary achievements for the real world, simply because it effectively offers opportunities to convey our design ideas into reality productivity. DSP algorithms are implemented by DSP systems and are then widely applied in various circumstances. Their main applications include biomedicine, digital image processing, speech/audio signal processing, instrumentation/control, military/surveillance, radar, telecommunications, seismology, consumer applications etc (Gaydecki, 2004, Ifeachor and Jervis, 2001).

Traditionally, most DSP systems consist of either simulation software or hardware equipment. However, advantages of numerous DSP systems may fade perhaps because of restriction of operating environments like physical location, or delay of off-line implementation.

Take some well-known DSP development tools for example, MATLAB and Simulink (offered by MathWorks) provide wide-ranging tools for signal acquisition, analysis, processing, visualization and algorithm development. Design ideas for real-time DSP systems can be simulated and validated by means of built-in libraries including finite impulse response (FIR) and infinite impulse response (IIR) digital filter design, analogue filter design, transforms, windowing algorithms, spectral/time-series analysis, statistical signal processing operations and so on (MathWorks, 2010a). MATLAB and Simulink are capable of speeding up real-time algorithm evaluation and system verification tasks before implementing hardware testing and final production software. Rapidly developing algorithms and exploring DSP system design ideas by performing simulation and verification are the advantages of the software. However, workspace for MATLAB and Simulink must be manipulated on computers unexceptionally, even with special system facilities such as processor speed, hard disk space, RAM (MathWorks, 2010b). From an embedded product development point of view, by using this software only, there must be a lengthy implementation from early system verification to final product.

One of the industry-leading DSP system development platforms is the LabVIEW DSP module offered by National Instruments. LabVIEW is a block diagram-based graphical development environment with the modularity of a programming language designed specifically for test and measurement applications. LabVIEW programs perform as intuitive virtual instruments, which abstract the implementation details and facilitate rapid designing and deploying DSP application prototyping, so as to quickly build DSP systems and simulate online or offline systems (National Instruments, 2010a). Again, this advanced software must be operated on computers, which signifies that all application efficiency and performance only emerge in existence of computers and the software.

In fact, there are some products like embedded technology platforms NI SPEEDY-33 and Texas Instruments C6711/ C6713/ C6716 DSP Starter Kits. Interfacing these peripherals, the LabVIEW DSP module is provided with stand-alone application development capability. By using the LabVIEW DSP module, the target boards can implement DSP algorithms in a stand-alone manner. However, the sizes of the hardware boards (NI SPEEDY-33 PCB dimensions: 8.89 cm × 12.7 cm × 1.91 cm) may not fit with small-size portable embedded applications (National Instruments, 2010b). Those tools might be more acceptable in an experimental learning environment but not in the application of telemedicine.

Briefly, both the hardware and software of typical DSP system applications are usually implemented at the point of signal acquisition. That means they are restricted by physical location to, for example, laboratories or factories. In other words, the DSP simulation software and hardware equipment have to be controlled locally. Another impediment of common DSP system is that if the signal acquisition from a target is off-line, data manipulation cannot be simultaneous. It can be a critical issue for some time-sensitive applications such as clinical monitoring.

Due to the fact that much demand for DSP system applications are now concerned with real-time remote control and portable devices, the software and hardware discussed above are not suitable for the fast growing market of wireless and remote DSP system applications, such as telemedicine system. Current industrial developments demand distributed network and remote capability for on-line control and off-presence monitoring (Gallardo et al., 2006). With the explosion of the Internet, the developments in computing and networking applications have opened an innovative way to develop remote control and real-time DSP system communication. Therefore, Internet controlled applications are becoming a necessity rather than an option.

To a certain extent, Internet controlled solutions may possibly resolve the impediments of restriction of operating environments and delay of off-line implementation in DSP system applications. This improvement originates from the Internet-based real-time DSP laboratory. For instance, engineers from different companies or students from remote institutions may wish to gain up-to-date DSP experience, but they may not be able to be present in the laboratories in which the necessary DSP software and hardware are located. This is made possible via remote Internet access. DSP hardware laboratories may then be developed using either fixed-point or floating-point DSP processors to support various applications, like filtering, spectrum analysis and communication applications. Thus, there is much research currently invested in the designs like the online real-time DSP laboratory. Take one successfully tested project for example, *'An Online Web-Based Real Time Digital Signal Processing Course'*, a representative online real-time embedded signal processing laboratory comprises a remote computer, a DSP evaluation module board, compatible software, some peripherals such as function generator and oscilloscope. The online users can use any local computer to access the remote computer with LabVIEW using network protocols. LabVIEW controls the function generator over a graphical user interface (GUI) on the remote computer and configures the waveform parameters which are fed to an analogue-to-digital converter (ADC). After the DSP evaluation module board processes the signals, the outputs of these are displayed from the oscilloscope to another LabVIEW GUI on the remote computer. The latter LabVIEW GUI controls the oscilloscope's features and settings (Ferzli and Karam, 2006). This laboratory describes an efficient way to access software and hardware remotely over network, which provides an operable mode to develop the Internet-based real-time DSP system.

Consequently, interest in remote real-time DSP systems has been growing immensely, accompanying the development of the network and communication technologies. In the meantime, due to the growing demand for clinical monitoring, the increasing popularity of remote instruments has had a profound impact on health-care and clinical medicine. Telemedicine is now a driving technology for modern healthcare delivery and provision.

In the last decade, the problem of controlling a real-time DSP system through an appropriate communication medium has been extensively researched and many approaches have been proposed (Kin Fun and Wei, 2011, Bansal et al., 2009). Especially, many fields of study in the applications of telemedicine are impressive. Different telemedicine service systems have been proposed or established. A typical prototype system is composed of a data acquisition module, a signal processing module and a PC with network connectivity as a transmission gateway (Hernandez et al., 2001). Often, signal processing and client-server communications are implemented on the PC by software like LabVIEW or MATLAB/Simulink. In this configuration, the computer on the patient side is indispensable (Reske and Moussavi, 2002, Garcia et al., 2002). Again, based on this system construction, physical location restriction is inescapable.

Recently, telemedicine systems have improved with development of some new transmission media. Much research has been carried out on the development of techniques for radio frequency (RF) telemetry of biomedical signals. With mobility, telephone networks, this provides patients better efficiency and effectiveness of the transmission system (Engin et al., 2005). A more advanced mobile phone platform personal digital assistant (PDA) with mobile General Packet Radio Service (GPRS) communication makes the implementation portable (García-Sáez et al., 2009, García et al., 2007). Normally Bluetooth technology is applied for wireless connection between the clinical signal acquisition module and the PDA. The deficiencies of these developments at present are the limited transmitting rate, non-universality and instability of the GPRS network service as well as the Bluetooth's short transmission range.

Recent developments in wireless communications have enabled telemedicine systems to expand in relation to benefits, applications and services. The evolving wireless telemedicine systems provide further flexibility, wider coverage and better healthcare delivery. A wireless local area network (WLAN) based on the IEEE 802.11 standard enable a number of access points with a range of 200-400 metres. Also, a computer is employed as a server (Pandian et al., 2007).

Many current developments share an essential common ground, that is, the personal computer or smart phone/PDA acts as a transmission gateway which provides facility to store and forward patient's signals. Additionally, the solutions are dependent on the professional software. Up to now, current telemedicine systems with network connectivity meant complicated system/network architectures with multiple-combined devices, cabling, and custom software. The merits and disadvantages of this architecture are distinct. The mobility of the patient and the scope of the applications are bounded by the PC and the network service provider. Moreover, the utilized hardware and software are often costly and require expert involvement.

To put it briefly, locating a PC/smart phone/PDA as a server or gateway might be inconvenient, even unfeasible, in most of the engineering application conditions. The usage of professional software like LabVIEW/MATLAB/ Simulink is costly and PC-based on patient site. The GSM/GPRS network service with the smart phone/PDA for telemedicine applications is less stable and much slower than the Internet. Particularly, all issues listed above hamper the widespread application of telemedicine.

Any existing problem of sufficient gravity is always an impetus for technological development. For live performance, if a portable platform is exploited as a bridge between users and systems by means of the ubiquity of the Internet, the real-time DSP systems for telemedicine can be disengaged from the bondage of limited applications.

To this purpose, a novel Internet-based real-time DSP system for telemedicine has been constructed, developed with the utilization of new technologies and it is described in this thesis. **The innovative point of this design** is concept of an integrated portable telemedicine system within a single platform, with neither the need for any additional cumbersome equipment/computer/laptop/smart phone/PDA and proprietary software on the patient side, nor any expensive dedicated server centre. The front-end hardware comprises biomedical sensors, a stand-alone DSP system, multi-function microcontrollers and their peripherals. This design provides real-time data acquisition, processing, distribution, visualization, storage and reappearance over the Internet. The full functionality of the system is visible to the Internet-based users who are able to view signals from the wearer's body, provided by the portable hardware. The system could have extensive applications in the educational, industrial, medical and entertainment fields where signals need to be acquired and monitored.

The rest of this thesis is organized as follows.

In Chapter 2, a detail analysis of the research related literatures is provided. This chapter begins with a review of the current background to telemedicine. Some DSP techniques in various clinical applications are listed, and the existing modalities of those DSP systems are presented. In the rest of the chapter, the important technologies in telemedicine are highlighted.

Chapter 3 contains an exhaustive description of the supporting theory and principles. The profiles of the target medical signals such as electrocardiograph (ECG) are examined first. Then the data acquisition techniques are described in detail. This is then followed by signal processing algorithm design that includes miscellaneous filter types. Diverse Internet protocols that support data transmission are included at the end.

Chapter 4 discusses the structure of the system platform including both hardware and firmware development. With an emphasis on innovative design, an overview of the original system design reveals how the new solution resolves the existing issues that are discussed in Chapter 1 and 2. All sub-modules of the system are described here and their functions are discussed. The description follows the progress of developing the entire system, from the front-end to the back-end.

The key experimental results are illustrated and analysed in Chapter 5. The interpretation of the associated discussions is given to each the experimental results in every module or sub-module.

Finally, further research and the conclusion are covered in Chapter 6. The main conclusion of the research programme encapsulates what have been achieved in this work. The proposals for this system improvement give some suggestions to the future developments in a systematic manner.

CHAPTER 2

Literature Review

2.1 Background

In clinical medicine and health care, conventional processes for gathering patients' clinical information are based on bedside implementation. Figure 2-1 portrays a flowchart of the traditional process of clinical signal collection, delivery and processing (Rolim et al., 2010). The disadvantages are obvious: increasing hospitalisation; increasing cost and time; critical scarcity in human resources; delay to the process; impacting on information delivery; and so forth. As a result, the traditional methods prevent real-time monitoring and remote access capability.

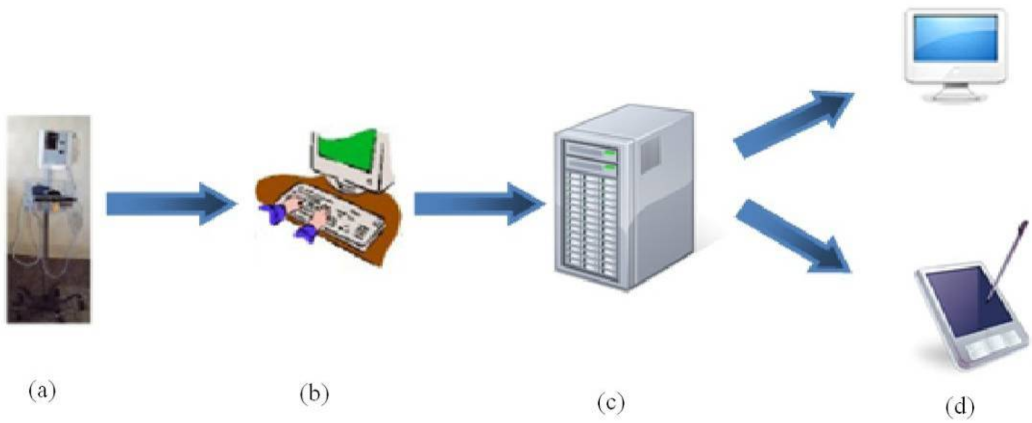


Figure 2-1 Traditional process of clinical signal collection, processing and delivery. (a) Collect data at bedside (b) Enter data in computer (c) Store data in database (d) Access data through interface applications.

The drawbacks and inconveniences discussed above can incur diagnosing delay, or more seriously treatment missing out. For instance, the clinical ECG data of the heart disease patient is collected and recorded in a ward of a hospital or at home and then is analysed by the doctors. There is inevitable latency or delay during patient signal collection and recording to the database. However, the monitoring and treatment of the heart patients must be implemented live. Otherwise, paroxysmal state and updated healthcare cannot be controlled in time.

In keeping with what was stated in the last chapter, if a remote real-time DSP system platform is bridged between users and systems by the use of a distribution network, an unprecedented revolution in health care will walk into our general lives. During the last decade, the development of computing and networking applications has opened a new way to understand control and instrumentation systems. Information and communication technologies (ICT) make pervasive use of telecommunication networks, especially the Internet, to design remote and virtual applications in the field of control, electronic instrumentation, DSP applications, etc. It is irreversible that the progress in clinical medicine and health care require integration with ICT methods and resources (Gallardo et al., 2006, Mazurek and Stroinski, 2010).

Telemedicine is one of the most significant achievements benefiting from ICT propagation. The term, telemedicine, is described as 'the provision of health care services, clinical information, and education over a distance using telecommunication technology'. The term exists long before the Internet (Maheu et al., 2001). Other expression of telemedicine may be argued to depict the term such as Telehealth or E-health. In this thesis, the spotlight is the development of patient clinical signal acquisition, processing and transmission so that the different expression of the terms becomes less emphasis. Furthermore, this work focuses on the technological development in order to enhance the service rather than the medical treatment itself. Therefore, the term telemedicine is borrowed and used here universally for all forms of electronic clinical medicine or health care delivered through networks.

Telemedicine evolves from store-and-forward to remote monitoring, from cabled to wireless. Store-and-forward methodologies acquire biomedical signal and then transmit them at a time convenient for off-line assessment; remote methods often involve real-time processing. Telemedicine proffers a promising pathway to enhance health care services and is beneficial for region-restricted people. Sharing patients' real-time clinical biomedical signals between clinicians can efficiently improve diagnosis and treatments.

'Tele' from the term 'telemedicine' dictates telecommunications or telemetry technologies in the applications of clinical biomedicine. The relational technological development that accelerates telemedicine involves three fields: biomedicine, signal processing and transmission.

DSP algorithms perform an important function to characterise biomedical signals from biomedical sources. It is ubiquitous in this discipline that signal contamination arises by artefacts such as respiration, body or muscle movements and electrical noise from the environment and electrical or electronic devices. The useful features of biomedical signals are often buried in interfering noise because the signals are weak, typically in mV or μV. So signal enhancement and feature extraction are probably the most frequently used functions of DSP applications in biomedicine. Those functions can minimise the distortion of the signal and characterise the signals of clinical interest (Wu et al., 1997, Outram et al., 1995). Many applications of DSP algorithms in biomedicine are extensively employed in the analysis of ECG, electroencephalograph (EEG), body temperature, respiration, and other vital signals.

Nevertheless, DSP algorithms eventually are implemented by hardware. DSP system hardware represents the embodiment of the applications of the algorithms. There are several suitable modalities in clinical medicine: implanted, wearable, portable, and environmental. They undoubtedly share a unique intention — better service including the patient condition, preferences, comfort, safety and cost (Varshney, 2009). These technologies normally employ a variety of particular sensors in different applications in order to collect typical information from the patients. An assortment of devices or software tools processes the data to extract the information of clinical interest. Finally, the signals that were processed by means of DSP systems are transmitted by a range of communication approaches and eventually reach clinicians.

Transmitting processed clinical signals is driven by the rapid growth of ICT, especially wireless/telecommunication technologies. Wireless/telecommunication technologies play a key role within telemedicine in an attempt to improve patient care; reduce costs, streamline processes, help with regulatory compliance, and provide many other benefits (Alasaarela et al., 2009). In general, the transmission gateways are utilized with the compatible networks to deliver the signals. For instance, a PC or laptop is commonly used in the environment of either a local area network (LAN) or WLAN so as to transmit the signals to the universal Internet. Take a more prevalent solution as an example: a smart phone with Bluetooth capability can broadcast the medical information through a commercial cellular/3G network. Moreover, an ad hoc wireless network is also an alternative in such applications.

The most up-to-date and relevant telemedicine systems are discussed and analysed in this chapter. All of the telemedicine projects share the same nature, that is, multidisciplinarity. According to this, based on relevance and different fields, this literature review presents three major sub-categories: DSP technologies in the clinical applications are discussed in the beginning. The second focus on the methods of clinical signal acquisition. The third pays attention to the technologies in telemedicine respecting mechanisms of transmission.

2.2 Signal processing in clinical applications

In telemedicine, physical biomedical signals are captured by a variety of sensors for monitoring and diagnoses. The information with medical interest from the body is extracted and converted from diverse energy forms like bio-potential, temperature, pressure and so forth into an electrical output. Usually the raw output of the measurement is of low amplitude and degraded by noise so that, in this raw form, it is unsuitable for display and interpretation. Hence, amplification and conditioning are necessary prerequisites prior to further processing. To recover the signals from noise, signal processing algorithms are used in many biomedical applications and various signal processing systems have addressed a widespread of clinical applications, especially in telemedicine. In this work, ECG signal acquisition is the main application.

To avoid unnecessary theoretical description, more profound principle details will be confined to Chapter 3, which considers algorithm theory in depth. This section focuses on the current developments or research areas.

The ECG is one of the most vital physiological signals, which directly reflects the heart's condition. ECG sensors noninvasively measure waveforms of cardiac electrical potential, which manifest as voltage differences generated during heart contraction. In general, two electrodes acting as a differential pair are used to detect a cardiac signal. The output from each pair is defined as a lead. In the context of ECG, the term 'lead' may either refer to a cable in the common parlance or mean the output of one electrode pair. There are different characteristic types of ECG sensors, such as integrating multi-leads or single-lead; the outputs are fed to an acquisition and processing module. The 12-lead ECG is the standard one which yields 12 views of the heart's electrical activities. Ten electrodes are usually employed to generate 12 channels. Ten electrodes and their placement are enumerated as follows (Conover, 1992, Welchallyn., 2010).

Table 2-1 Ten electrodes and their placement

Label	Electrode Placement
RA	Far down on the right arm, or below the right clavicle
LA	In the corresponding position of RA’s, but on the left arm
RL	Far down on the right leg, or above the right hip
LL	In the corresponding position of RL’s, but on the left leg
V_1	In the fourth inter-costal space at the right border of the sternum
V_2	In the fourth inter-costal space at the left border of the sternum
V_3	Between location V_2 and V_4
V_4	In the fifth inter-costal space at the mid-clavicular line
V_5	Horizontally even with V4 at the anterior axillary line
V_6	Horizontally even with V4 and V_5 at the mid-axillary line

The limbs can be regarded as line conduction or lead extension. Therefore the electrodes in the different positions along the limbs virtually result the effective outputs. The following Figure 2-2 shows the layout of the ECG electrodes as quoted from the American heart association (AHA) (Welchallyn., 2010).

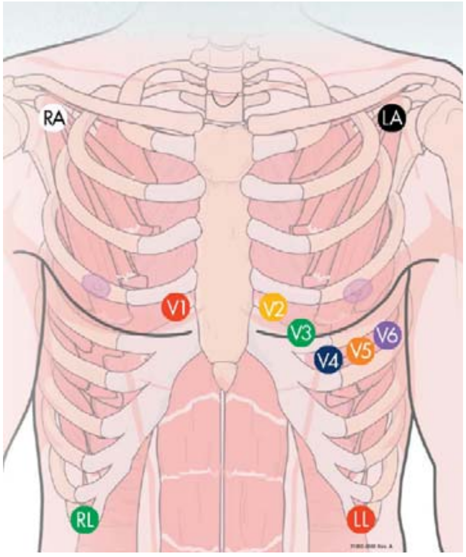


Figure 2-2 AHA diagnostic ECG electrode placement (Welchallyn., 2010)

These ten electrodes in the different positions of the body generate twelve channel signals by particular permutations and combinations. The twelve channels are grouped as limb leads – Lead I, II, III; augmented limb leads – Lead augmented vector right (aVR), Lead augmented vector left (aVL), Lead augmented vector foot (aVF); precordial leads — V_1, V_2, V_3, V_4, V_5 and V_6 (Prineas. et al., 2010).

A typical 12-lead ECG shows a short segment of the 12 outputs which consist of four columns by three rows. It is difficult for a clinician to analyse heart rhythm by only a few heart beats. With the purpose of rhythm analysis, a normal trace of 12-lead ECG is usually used as illustrated in Figure 2-3 (Vecht et al., 2009).

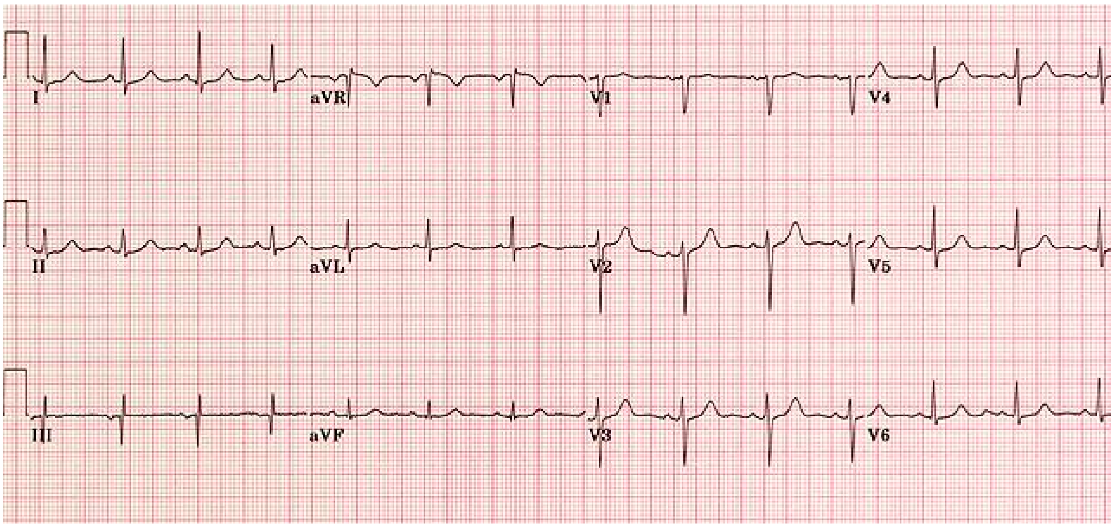


Figure 2-3 Normal trace of 12-lead ECG (Vecht et al., 2009)

Commonly, prevalent medical monitoring instrumentations such as the Holter monitor use a 3-lead ECG rather than the traditional full 12-lead ECG record data. The standard 3-lead ECG is applied for continuously observing and quick diagnosis, locating three electrodes on the vertices points of what is known as ‘Einthoven’s Triangle’. One of Lead I, II or III is measured as a single signal trace, which is generally sufficient to monitor life threatening disruption (Iaizzo, 2009). Long-term or ambulatory monitoring methods also apply either one-lead or two-lead ECG.

Traditional bio-potential electrodes/sensors use silver/silver chloride (Ag/AgCl) electrodes placed on the human body surface to measure electrical potentials. The ionic current flow in the human body is measured and converted to electron current flow by the electrodes as transducer. Between the living tissue/skin and the electrodes, an electrolyte solution of silver chloride with gel-adhesive is coated at the centre of the electrode to form a conductive path. A chemical reaction occurs at the electrical interface and makes the current pass between the electrolyte and the electrode. The electrolyte jelly is on one side of the electrode coming into contact with living tissue/skin. Another side of the electrode is conductive metal linked to a lead which feeds the signal to the signal conditioning unit (Lee. and Kruse., 2008).

In practice, many challenges are unavoidable when using this kind of standard Ag/AgCl electrode with electrolyte gel-adhesive. First of all, because of dehydration associated with the electrolyte gel, the service life is effectively a few days. Accordingly long-term physiological monitoring during the period of the Ag/AgCl electrode's operating life is not practicable. Secondly, dehydration leads to the electrolyte impedance increase and DC offset increase, so that additional artefact and noise are generated. This causes further challenge to the signal process and display systems. Problems from mutative impedance in the electrolyte also issue from different objects – dry or old skin, patient's age, gender and ethnicity. What is more, the gel can breed skin irritations and bacterial growth (Searle and Kirkup, 2000, Lee. and Kruse., 2008, Karilainen. et al., 2005).

To circumvent the side-effect of the use of electrolyte gel-adhesive, there has been increasing interest in dry textile/ fabric ECG electrodes. Some research efforts focus on dry electrodes formed from conductive yarns and filaments (Mestrovic et al., 2007). Dry electrodes consist of a benign metal used with no electrolyte between the electrode and skin. Those challenges discussed above can be eliminated by utilizing the new electrodes (Assambo and Burke, 2009, Searle and Kirkup, 2000, Pola and Vanhala, 2007). However, dry electrodes suffer from their own serious predicament like loss of contact and artefact. It is been stated by some researchers that the performance is similar between the dry textile/fabric electrodes and the electrolyte based electrodes, and that there is no medical and diagnostic information lost (Enzo Pasquale et al., 2005). New methods for the clinical ECG monitoring have taken the form of intelligent clothing as wearable measurement platforms, in which dry textile/fabric electrodes are integrated (Lee. et al., 2009). More detailed wearable modality applications will be elaborated in the later sub-chapter.

The inherent character of the ECG signals which are detected from an assortment of electrodes is of low voltage analogue potentials ranging from 0.1 mV to 4 mV (Yazicioglu et al., 2007). Inevitably, these bio-potential signals are submerged within a mixture of noise. Common mode noise is the major interference which is induced on the human body because of the coupling capacitances between itself and the mains supply. The amplitude of the common mode noise generally overrides the ECG signal. Consequently, to acquire and process valuable signals, a precision analogue instrumentation amplifier is employed to regulate the outputs which are fed from the leads at the input stage of the system. The instrumentation amplifier is required to implement the functionalities of attenuating the common mode noise and amplifying the ECG signal in a differential mode. The selected amplifiers are provided with high differential gain, low common mode gain, high common mode rejection ratio (CMRR) and high input impedance (Griffiths. et al., 2007, Chaudhuri et al., 2009). An example of an ECG-ambulatory/Holter monitor AC-coupled functional diagram is excerpted below (Analog Devices, 2010a) in Figure 2-4.

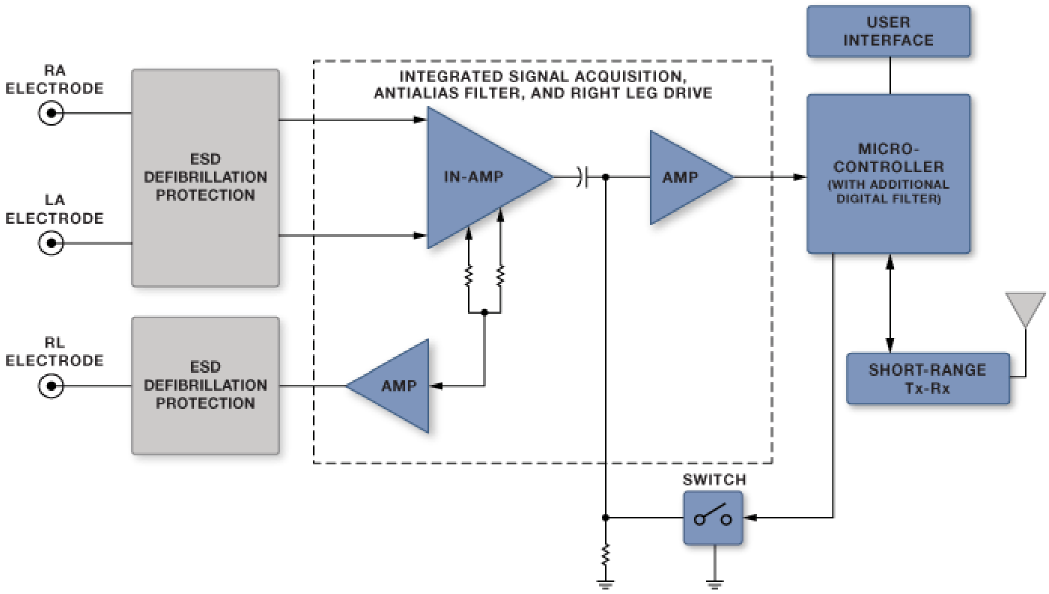


Figure 2-4 ECG-ambulatory/Holter monitor AC-coupled functional diagram (Analog Devices, 2010a)

For multi-channel simultaneous physiological signal acquisition, each channel requires one instrumentation amplifier. If synchronization is not critical, a single unit can be applied to amplify all channels by using a multiplexing scheme.

After amplification, the ECG signals are typically still contaminated by various sources of noise. Filtering the noise to recover the ECG prototype is the kernel of signal processing, a precondition of which is to identify the frequency spread of the electrode outputs. The mixed low frequency noise arises from baseline wander, respiration and electrode surface movement artefact against the skin. Other interference originating from the radiated electromagnetic environment, electromyography (EMG) muscle artefact and power line needs to be eliminated (Kligfield. et al., 2007, Blanco-Velasco et al., 2008).

High pass filter with cut-off frequency at 0.5 Hz is widely used in many ECG monitors. Nevertheless, it can result in artefact deviation of the ST-segment. A 0.05 Hz low-frequency cut-off is recommended by the American heart association for diagnostic ECG. In this case, digital filters have superior performances over traditional analogue filter methods in the usage of the high pass filter. To measure the ECG signals with acceptable diagnostic accuracy, the upper cut-off frequency is 150 Hz for adults, adolescents and children, and up to 250 Hz for infants. It is comprehensively recognised that the majority of the diagnostic information is contained below 100 Hz in adults for the visual interpretation of the ECG signals (Rijnbeek et al., 2001, Griffiths. et al., 2007). Generally, a notch filter at 50/60 Hz is widely used to suppress the power line/mains frequency interference.

Nearly all ECG instruments convert the analogue signals to digital form for further processing, transmission or storage. According to the Nyquist theorem, the ADC sampling rate must be at least double the highest expected frequency which the signal contains. With the aim of preserving the higher frequency components of the ECG signals and further processing, the initial sampling rate is often many times faster than the expected one, which is called oversampling (Kligfield. et al., 2007, Goldberger. and Ng., 2010). Amplification and filtering also need to meet the amplitude and frequency range of ADC.

The processed ECG signals normally can be transmitted through transmission gateways like a computer or smart phone within diverse networks. Alternatively, they can be stored over cable or wirelessly to memory card or database, as discussed previously.

Many current DSP systems for ECG have been developed to cope with a range of clinical and technical challenges so as to obtain diagnostic signals and real-time processing. Capua et al. (2009) have proposed a solution for ECG signal processing. A differential amplifier

AD8220 from Analog Devices™ offers high CMRR to reject the common mode signals like baseline wander interference and electromagnetic interference (EMI). It is employed to realise a high-pass analogue filter at a cut-off frequency of 0.033 Hz to remove DC offset and frequency artefacts. A low-pass active filter is designed at a cut-off frequency of 160 Hz by using an AD8616. Both amplifiers implement the functionality of signal amplification. The circuit also includes an OP2177 operational amplifier for buffering, and a resistor at the output of the OP2177 to isolate fault currents to protect the patient's body. A 12-bit dsPIC30F3014 from Microchip™ is used, which incorporates an ADC. The digitized signals are then fed to a PDA via a serial port. A programme on the PDA implements a band-pass FIR digital filter at frequency range between 0.05 Hz to 150 Hz as well as a notch digital filter to remove 50 Hz. The photograph is excerpted to show the system (Capua. et al., 2009) in Figure 2-5.

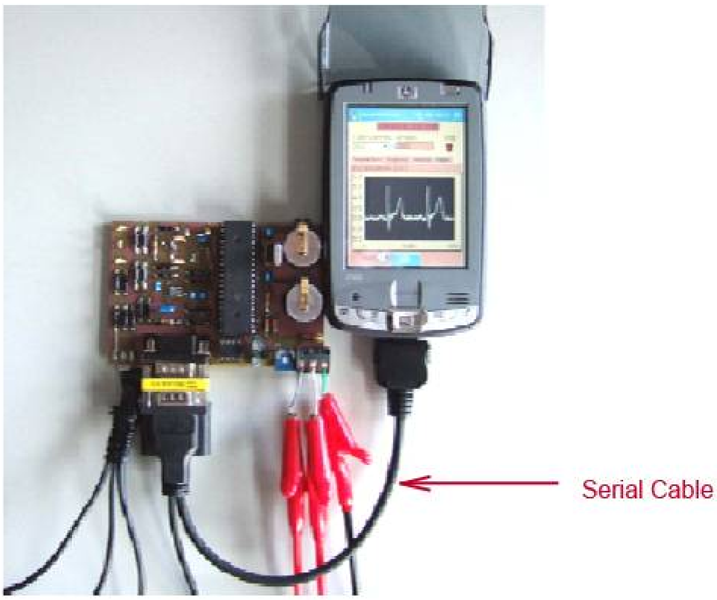


Figure 2-5 The ECG measurement system (Capua. et al., 2009)

The analogue circuit details and the processing in the PDA are described in the paper. Even though the device is portable based on the PDA, a bunch of cables have to be used, which result in suppressing its capability and practicability. Besides, the system is deficient in universal distribution and only offers patient self-monitoring functions in the PDA. The acquired ECG data is simply stored in the PDA or is printed out through Bluetooth connection. System amelioration is necessary for telemedicine performance.

Other prototype ECG system was developed by Bansal et al (2009). This computer-based wireless system for online acquisition, monitoring and digital processing of ECG waveforms includes an analogue module, an FM transceiver pair and a computer which displays and filters the real-time ECG signals using MATLAB software. The functional block diagram is illustrated in Figure 2-6 (Bansal et al., 2009).

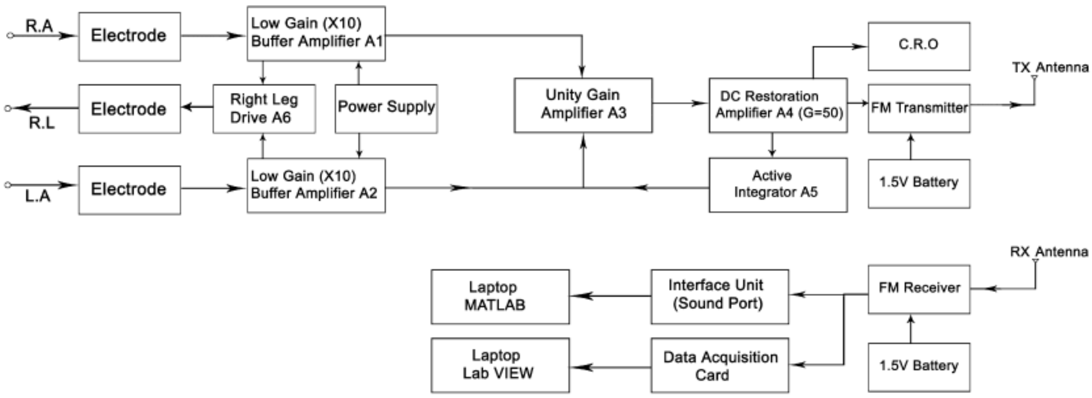


Figure 2-6 Functional block diagram of the prototype ECG system (Bansal et al., 2009)

The prototype system has three modules: a front-end amplifier, wireless FM transceiver and interface with output display device and digital filters for noise removal. In the analogue section, traditional Ag/AgCl electrodes with electrolytic gel sense the 1 mV ECG signals. The low amplitude bio-potential signals are amplified in multiple stages by an operational amplifier TL084C which has a CMRR of 86 dB and adjustable gain up to 500 in cascade. The amplifier circuit executes the functions of buffering, unity gain amplifying, DC restoration and reducing the CMRR by means of a driven right leg circuit. Active analogue filtering is also included. The processed signals are sent to a computer for further processing by an FM transmitter receiver pair in the frequency range of 88-108 MHz. The received signals is fed to the audio port of the computer, and then visualised and processed by MATLAB. The flow chart of the MATLAB programme to filter the ECG signals is described in the paper (Bansal et al., 2009) as shown in Figure 2-7.

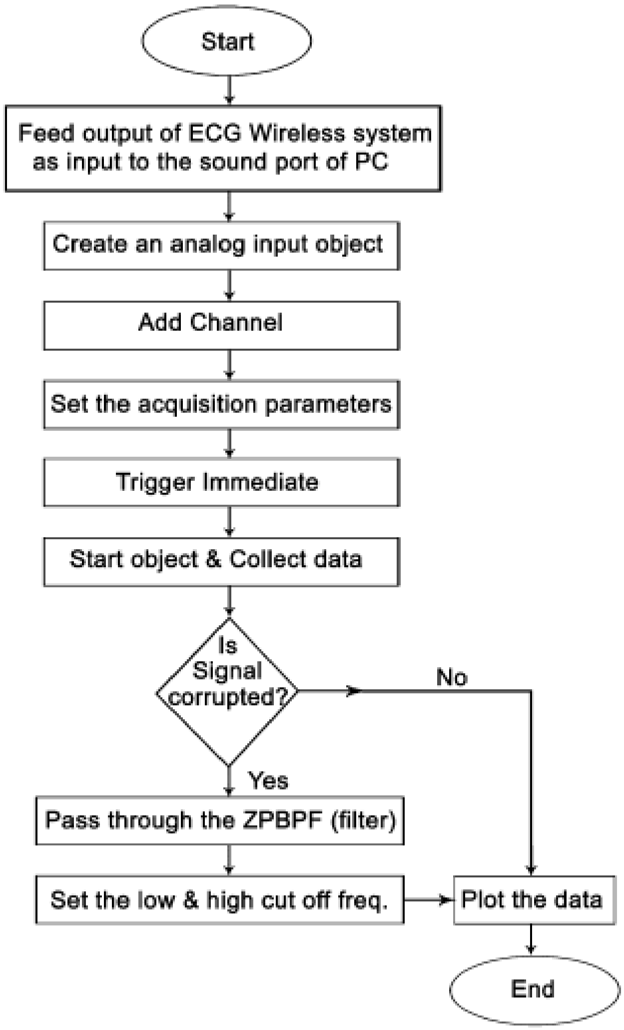


Figure 2-7 Flow chart of the MATLAB programme (Bansal et al., 2009)

In this MATLAB programme, a zero phase band pass filter and FIR digital filter are employed to effectively analyse and filter the ECG signals in real-time. Furthermore, a data acquisition card from National Instruments^{TM} as well as a LabVIEW solution for recording, analysing and displaying the ECG data is also proposed. Although the system can capture and display the ECG signals in real-time on the laptop, the usage of the laptop limits patient's activity.

In short, the bio-potential signals, like ECG measured by the ECG electrodes or sensors, are normally difficult to be detected for directly observing and monitoring. Additionally, the interference from various sources can obscure the medical signals. Thus, amplification is unavoidable to enlarge the amplitude of the signal, but concurrently the execution also amplifies the noise. On the whole, most solutions exploit analogue filter circuits after amplification and afterwards implement further signal processing in the software environments such as LabVIEW or MATLAB. It is noticeable that the deployment has to comprise computer or PDA. From the telemedicine point of view, these reviewed up-to-date systems cannot offer truly portable, highly-integrated ECG instrumentation with widespread distribution. Those designs without network connectivity are impossible to achieve globally accessible data and monitoring.

2.3 Methods of clinical signal acquisition

From the last section, it is apparent that the physical biomedical signals typically of low amplitude and subject to distortion because of interference. A variety of signal processing methods can applied to extract features of medical interest. To do so, there must be a precondition, that is, the measurement which transforms the biomedical energy to a usable electrical signal. This mission falls to a range of specialised sensors or electrodes. Varshney (2009) in his recent book characterises the suitable information and communications technologies for healthcare/telemedicine according to four categories: implanted, wearable, portable and environmental. This classification might not cover the entire range of modalities, but it serves as a useful starting point. This section will introduce the current technologies for clinical signal acquisition, following this classification.

2.3.1 Computer-based implementations

Traditional clinical measurement or monitoring mechanisms in telemedicine, in either store-and-forward or real-time mode, are based on workstations, or bulky medical instruments or computers equipped with dedicated software. An assortment of cables connects and transmits the biomedical signals from various specialised sensors to the workstations. Generally, the preconditioning modules and the I/O interfaces (ADCs) are obligatory before feeding the signals to the workstations. The operation is uncomplicated and reliable on site or in the examination room. The computing facilities can offer smart monitoring during patient's diagnoses and treatment. The computing technologies are essential for the increasing e-health system.

Some examples of computer-based implementations like MATLAB/SIMULINK and LabVIEW/SPEEDY-33 were discussed in the last chapter. Another successful example is a computer-based wireless system for ECG waveforms capture and transmission. The ECG signals can be acquired and processed by a laptop either running MATLAB through its sound port, or alternatively running LabVIEW through an NI data acquisition (DAQ) card (Bansal et al., 2009). The functional block diagram of the system is illuminated in Figure 2-6.

Briefly, even if the computer-based telemedicine systems have certain advantages, yet these systems are equipped with many cables between the sensors and the workstations, which inescapably hamper patients' activity and comfort. This is a conspicuous obstruction for the evolution of telemedicine development.

2.3.2 Portability

Although the computer based systems were in previous years quite bulky and software-dependent, miniaturisation in recent years has enabled development of portable devices. The Holter device is now a standard clinical instrument used to continuously monitor various biomedical signals from patients for 24 hours and then to perform off-line recorded data analysis. It normally consists of electrodes, signal acquisition unit, battery supply unit and data storage unit (Jovanov. et al., 1999). The Holter devices are used merely to record clinical signals from ambulatory patients; processing and analysis are performed off-line on recorded data.

In the applications of telemedicine nowadays, new designs have incorporated the conventional Holter device and wireless transmission technologies. A Holter device for ECG continuous monitoring using wireless technology has been designed to improve the performance of the devices. A standard analogue front-end module which is composed of some amplifiers and filters pre-conditions the raw ECG signals. The modulated signals are digitised by an on-chip microcontroller incorporating an ADC and then transmitted through a universal asynchronous receiver/transmitter (UART) serial interface via Bluetooth. The signals can also be stored to external SRAM memory by this microcontroller. The block diagram of the Holter device is excerpted in Figure 2-8 (Led et al., 2004).

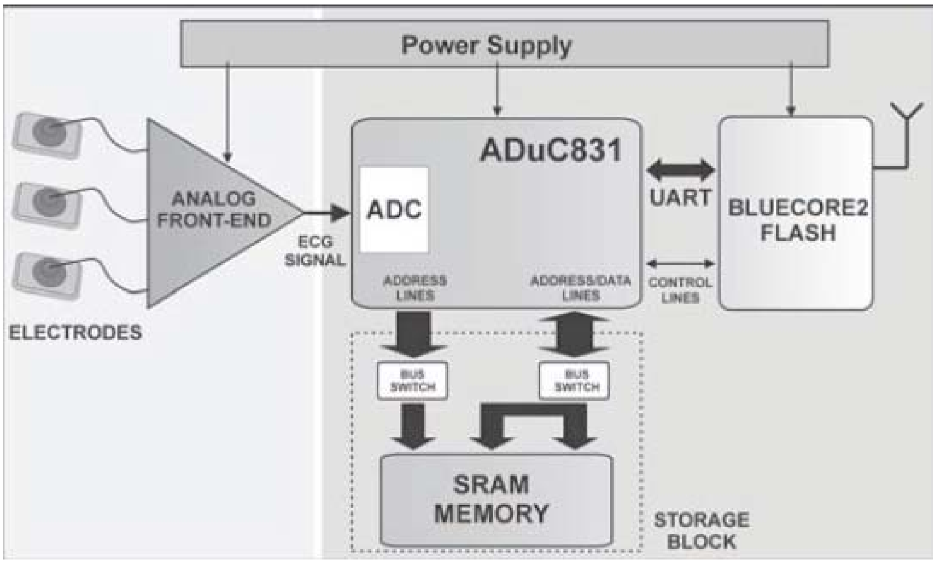


Figure 2-8 Block diagram of the Holter device (Led et al., 2004)

With the aim of accessing patient's monitored data remotely, a laptop, PDA or GPRS phone can be employed as a gateway to transmit the data to a database. By using Bluetooth technology, a BlueCore2-Flash chip operates as a wireless link between the Holter device and the gateways (Led et al., 2004).

Other miniaturised portable devices are also being developed but all of them unavoidably utilize a laptop/PDA/GPRS phone/smart phone as gateway (Gu-Young and Kee-Ho, 2006, Zhu and Wang, 2005, Capua. et al., 2009, Ashraf, 2009).

Without a doubt, portable solutions can offer patients the capability of ambulatory freedom and provide clinicians remote access to the data. However, the usage of such gateways is always a hindrance to universal connectivity and real-time operation. What is more, the gateways are normally costly and dependent on the bandwidth-limited GPRS service.

2.3.3 Wearability

Accompanying prevailing portable solutions for telemedicine systems, conventional sensors/electrodes with electrolyte gel as well as conduction leads are widely used to measure the physiological signals and still find common use in telemedicine. Such front-end modules obviously impede the evolution of this technology. The disadvantages of the electrolyte gel's usage and the inconvenience of the cables for patients have been discussed in section 2.2.

To cope with these impediments, the non-invasive monitoring of biomedical signals using new technology, dry wearable sensors, is an emerging and rapidly growing trend. Typically, dry belt-type or textile sensors/electrodes and the leads are integrated into a vest for the continuous monitoring of various biomedical signals. The sensors/electrodes are distributed at specific body locations to accurately detect the biomedical parameters (Choi and Jiang, 2006, Pandian et al., 2008, Wenxi et al., 2005, Chulsung et al., 2006). The 'Smart Vest' is an example of a wearable physiological monitoring system, which is proposed to integrate an array of sensors to acquire ECG, photoplethysmogram (PPG), body temperature, blood pressure, galvanic skin response (GSR) and heart rate (HR) signals. The dry sensors including Silicon rubber with pure silver yarn in the ECG sensors as well as the conduction wires which are stitched into the fabric, based on two belts of the vest. The following illustration shows the excerpted diagrammatic sketch (Pandian et al., 2008) in Figure 2-9.

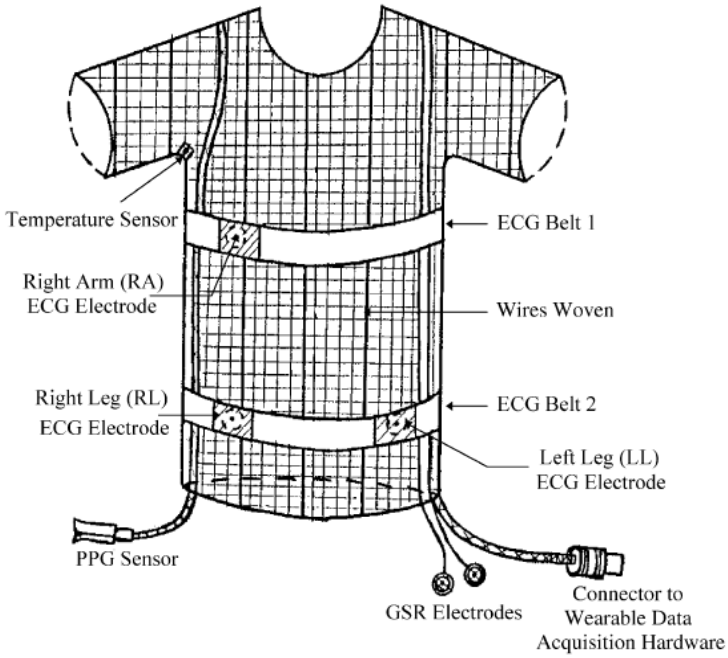


Figure 2-9 Diagrammatic sketch of 'Smart Vest' (Pandian et al., 2008)

Connected to the output leads of the vest, a wearable data acquisition hardware unit implements signal conditioning, digitisation, processing and wireless transmission. The analogue ECG signals are amplified by a two-stage differential amplifier with gains of 10 and 100, i.e. a combined gain of 1000. High pass and low pass filters limit the pass band within 0.5-100 Hz after the first and the second stage amplification respectively. Power line interference is suppressed by a 50 Hz notch filter. The processed signals are conditioned by a level shifter circuit to meet the ADC requirement. After that, the biomedical signals are dispatched to the ADC of a 16-bit microcontroller (dsPIC30f6014) and transmitted to a remote laptop as monitoring station via a wireless transceiver module for further processing and display (Pandian et al., 2008). This proposal represents an initial effort to develop a wearable clinical signal acquisition system by using dry sensors which are simply inserted in the smart vest.

Evolution and advance of miniaturisation in the technology has taken place not only with the portable wireless systems but also with the front-end sensors/electrodes, for example, the tiny wearable ‘QUASAR’ ECG sensor and the ‘Eco’, a wireless sensor node. The ‘QUASAR’ sensor integrates capacitive-coupled electrodes and a signal conditioning circuit, with a size of 15 mm × 3.8 mm, weighting of 5 g (Chulsung et al., 2006). The ultra-compact wireless sensor node ‘Eco’ operates as a signal acquisition module and a wireless transmission module (Chulsung and Chou, 2006).

With the purpose of designing comfortable, wearable and washable smart garment system, some innovative textile/fabric sensors/electrodes represent an important development in telemedicine. The typical sensor/electrode incorporates transduction functionality onto a textile substrate, based on rubber-carbon-coated thread technology and weaving topology (Yoo et al., 2009, Rienzo et al., 2005, Enzo Pasquale et al., 2005, Mestrovic et al., 2007, Pacelli et al., 2006).

Indeed, experimental results show that the textile/fabric sensors/electrodes may monitor vital physiological signals, to a certain extent, without loss of essential signal integrity. However, it is worth noting that the developments of textile/fabric sensors/electrodes are still at a pilot stage and there are several existing issues in the context of the actual applications. Problems like loss of contact, reproducibility of positions of the sensors/electrodes and movement artefacts are still challenging the superiority of the wearable mechanism. Nevertheless, the wearable intelligent garment could be the most comfortable solution for telemedicine in the future development, providing efficient monitoring of clinical biomedical signals with minimal impact on the daily life of the wearer.

2.4 Technologies in telemedicine

Current and emerging developments in wireless solutions and communication networks which are integrated with biomedical sensor technologies are now having a considerable impact on medical/healthcare service. Much of the literature shows, as suggested above, that the research has focused on incorporation of clinical disciplines and multiform signal processing modes. What endows clinical monitoring systems with the functionality of telemedicine is the ubiquitous connectivity between patients and clinicians. To deploy the connectivity to the biomedical monitoring systems, diverse wireless solutions and assorted communication networks are applied according to diverse circumstances.

2.4.1 Wireless technologies in telemedicine

Wireless solutions incorporating new technologies provide significant possibilities for the advancement of clinical monitoring systems, which in turn can offer freedom to ambulatory patients whilst being continuously monitored. Wireless deployment also increases the coverage of telemedicine to extend to the area where the wireless connectivity exists. In order to facilitate patient mobility, various wireless technologies have been integrated into telemedicine systems. Almost all biomedical monitoring systems acquire vital physiological parameters by biomedical sensors/electrodes. Normally, a standard front-end analogue signal conditioning module processes the biomedical signals. Diverse wireless technologies, such as RFID, Bluetooth, ZigBee, WLAN etc., are employed to feed the detected signals from the conditioning module to the monitoring centre or the gateway.

Bluetooth

In many biomedical applications, a wireless personal area network (WPAN) such as Bluetooth is employed to replace the wired connection between the signal acquisition device and the transmission gateway. The Bluetooth interface is extensively available in devices like mobile phones, PDAs, PCs, laptops, printers and so forth. These devices can be potentially configured as control centres or transmission gateways. The Bluetooth profile offers compatibility among devices of different manufacturers and universality in the market today. Furthermore, Bluetooth technology is characterized by low power consumption and reasonable throughput. It operates at a secure unlicensed radio frequency, which is hence suitable for small-size, battery-operated clinical biomedical devices. Thus, Bluetooth as a short-range wireless communication standard is commonly adopted for front-end wireless solutions in telemedicine applications.

Bluetooth uses a radio technology named frequency hopping spread spectrum (FHSS) at 2.4 GHz. In this short-range radio frequency band, the basic networking unit, namely piconet, consists of a wireless user group. A piconet includes up to eight devices, one of which performs as a master and the others as slave(s). Multiple piconet can be overlapped each other to connect to more devices. Bluetooth is a standard communications protocol with three different transmission ranges depended on power consumption: approximate 100 m (100 mW), 10 m (2.5 mW), and 1 m (1 mW), respectively corresponding to class 1, 2 and 3.

In some portable systems for acquiring and monitoring ECG signals, a Bluetooth transceiver is employed to transfer the signals through a microcontroller to a remote PC for further processing in LabVIEW (Griffiths. et al., 2007, Kang-Ming and Shing-Hong, 2009). Similarly, a real-time wearable system named HealthGear designed by Microsoft utilizes a cell phone to store and analyse the patient's blood oxygen level and pulse signals via Bluetooth (Oliver and Flores-Mangas, 2006).

Transmission gateways can offer the capability of remote access for physiological signals through diverse ubiquitous infrastructure networks. Combined with distributed networks, Bluetooth technology is used to support wireless connection between the signal acquisition module and transmission gateway. For example, Led et al propose a prototype using a single chip BlueCore2-Flash for wireless communication between a mobile Holter device and a laptop, PDA or GPRS mobile phone as gateway (Led et al., 2004). A mobile ECG and motion activity monitoring system HEART GUARD is based on a PC as a gateway to transfer data to the Internet by the use of Bluetooth (Korsakas et al., 2006). More current successful designs use smart phones or PDAs as gateways and communicate with GSM, GPRS or the advanced 3G network, accompanied with Bluetooth technology (Ashraf, 2009, Rasid and Woodward, 2005).

Indeed, the Bluetooth standard is appropriate for medical instrumentation development, which aims at convenience of usage and system reliability. In most cases, class 2 is employed with a short range of 10 m, which appreciably restricts the ambulatory patients' active region. In some systems, class 1 with 100 m range is used but this means that higher power consumption is inevitable. The limited transmission rate of Bluetooth signifies that it is only satisfactory for low frequency signal detection across a limited number of channels (Varshney, 2009).

ZigBee

ZigBee is a wireless communication protocol similar in concept to Bluetooth. The ZigBee protocol also operates in the 2.4 GHz unlicensed frequency band. ZigBee is a low-cost, low-power wireless mesh networking standard, which is intended to define a simpler and less expensive solution to Bluetooth. Although running in the maximum data rate only 250 Kbits/sec, ZigBee aims at large-scale and a large number of multi-users in an interoperable environment (Mulyadi et al., 2009, Fernandez-Lopez et al., 2010).

ZigBee technology is widely adopted to build front-end wireless body sensor networks (Chen et al., 2009, Camilo et al., 2009, Shuo-Jen et al., 2008, Xiao et al., 2008, Benocci et al., 2009). Xiao, H et al. in their paper compared three major wireless network technologies Wi-Fi, Bluetooth and ZigBee and proposed a ZigBee-based System-on-Chip (SoC) solution by using Texas Instruments' CC2430 (Xiao et al., 2008). The biomedical signals are dispatched to a mobile cellular network such as GSM or 3G by the ZigBee coordinator. Shuo-Jen, H et al. in their research proposed a new DSP platform incorporating a ZigBee-based wireless sensor network (Shuo-Jen et al., 2008). But this system uses a relay instruction by the ZigBee router to transmit the signals, which results in a lack of remote monitoring capability. A vital signs monitoring system called HM4All was explored to establish WSN by ZigBee protocol. A ZigBee coordinator and ZigBee to

Wi-Fi gateway (computer here) retransfer the signals to the Internet (Fernandez-Lopez et al., 2010). ZigBee can operate in active/sleep mode and the activated time can be as low as 15 milliseconds (compared to Bluetooth wake-up delay which is around three seconds) (Xiaoxin et al., 2010). It has low power consumption for the reason that the device can be designed to sleep most of the time.

Notwithstanding, the low transmission rate may be a stumbling block for ZigBee technology in applications which require the acquisition and transmission of high frequency biomedical signals.

WLAN

The WLAN was designed to provide short-range (up to 100 m) connection for both infrastructure (access point to the Internet) and ad-hoc (peer-to-peer) configurations. It operates in either the 2.4 GHz (widely used) or 5 GHz (limited use) unlicensed band according to the different version. The flexibility and mobility of WLAN are major attractions for wireless solutions.

WLAN in telemedicine applications has grown remarkably in the hospital setting. For example, a wireless physiological monitoring system was proposed to transfer continuously acquired patients' vital signs to a central management unit in the hospital, which integrates PDA and WLAN technologies. Due to the coverage radius of selected WLAN and regional confines, the distance between each access point was set less than 30 m, between which patients could roam (Yuan-Hsiang et al., 2004).

In terms of the drawback of mobility and coverage, WLAN may be only suitable for local services as a front-end infrastructure. In the applications of telemedicine systems, other widespread networks like the Internet or cellular network are required for back-end connection. WLAN also experiences issues such as signal attenuation or loss because of fading and dead spots. Furthermore, security is still a problem owing to difficulties associated with high-level encryption.

WiMAX

In order to provide medium distance (metropolitan coverage) wireless communications with broadband access, high data rate (about 40 Mbps, with updated versions expected offer up to 1 Gbps) and mobility, WiMAX technology has been recently introduced to the applications of telemedicine. WiMAX, an abbreviation of Worldwide Interoperability of microwave Access, is a telecommunications protocol which provides fixed and fully mobile Internet access in both indoor and outdoor environments. The WiMAX network can supply hundreds of users from a single WiMAX base station. Normally users are computers associated with application software at each node.

With the accessibility of broadband by WiMAX, quality imaging transmission and video conferencing in the ambulant status become possible in telemedicine. An economic and efficient telemedicine system called Emergency Wi-Medicine EWM was proposed. This system focuses on improving video conference performance providing communications between the hospital and ambulance or on-the-spot for diagnostic purposes (Lam et al., 2009). In the article (Niyato et al., 2007) a variety of WiMAX-based deployments in telemedicine scenarios are investigated. By using WiMAX technology, the telemedicine systems can gain considerable benefit from the high-rate broadband connectivity in a secure manner.

Compared to GSM, digital subscriber line (DSL) or fibre-optic network, the WiMAX network can be deployed rapidly at relatively low cost. However, it cannot yet provide ubiquitous availability since the WiMAX base stations and subscriber stations are presently few in number.

Other wireless technologies

Many other wireless technologies have been investigated as communication channels in the context of this research, such as satellites technology.

Satellite-based telemedicine systems have global broadcasting coverage. Although satellite communication systems offer the most mobility and flexibility, there are some drawbacks which confine its applications to exceptional circumstances. First, the high cost of these systems; second, the size and weight of the supported hardware is often prohibitive from the perspective of portability. Additionally, the communications can be affected by natural phenomenon like rain and snow (Varshney, 2009, Pandian et al., 2007).

Collaborative coexistence

Wireless technologies in telemedicine applications are vast, and only the significant ones are covered here. Nowadays, it is hard by only using simplex technology to make satisfaction for increasing quality of service (QoS).

According to individual characteristics such as coverage range, power consumption, transmission rate, capacity of multi-user systems and operating feasibilities etc, collaborative coexistence schemes have been proposed and explored on a large scale. A hybrid network has been proposed by (Mulyadi et al., 2009), which involves a wireless interface for a telemedicine system based on both Bluetooth and ZigBee technologies. Misic, J and Misic, V. B. in their research (Misic and Misic, 2010) discussed the possibility of using Bluetooth and WLAN technologies in an integrated healthcare monitoring system. Moreover, the WiMAX network is a promising solution to provide backhaul support for WLAN hotspots. Network backhaul is described as a direct connection between a base station and a network controller (Jelena et al., 2009, Krohn et al., 2007, Yan et al., 2010).

However, the inherent drawbacks like system compatibility to these technologies must be addressed during the development phase.

2.4.2 Cellular networks in telemedicine

Most wireless technologies can only operate at high transmission bit rates or over long distances but not both. Any one of these wireless technologies has its intrinsic characteristics which can affect the performance of telemedicine systems.

Recent developments in mobile telecommunication networks have brought new potential and exploitation to telemedicine systems. Mobile telecommunication networks can be employed as subsequent transmission networks following front-end wireless connections. With the breakthrough of wireless telecommunication technologies and deployment of cellular networks, mobile telemedicine services have become popular for providing prompt and effective patient monitoring. In general, the commercial mobile telecommunication networks have nationwide coverage. Patients can receive healthcare service as long as they roam within the region where the cellular networks cover. Generally, the transceivers of the wireless networks are cellular phones, smart phones or PDAs.

The fundamental architecture of mobile telecommunication networks are cellular, in which the base station coverage (the cell) is determined by the effective range of the radio antenna effective range. Commercial cellular networks include several generations; at present those operating are second generation (2G) and third generation (3G) and imminent fourth generation (4G).

There are two versions of 2G networks. They are TDMA-based (time division multiple access), e.g. global system for mobile communications (GSM), and CDMA-based (code division multiple access). GSM was primarily designed to propagate voice massages. The GSM-based 2G network represented the majority (80.5%) of the cellular phone users (3.7

billion) in 2008. (Tahat, 2009) Almost all 2G network applications in telemedicine, more specifically GSM-based telemedicine systems, employ the short message service (SMS) to deliver alerts respecting a patient's vital signs to doctors (Kale et al., 2007, Tahat, 2009, Sukanesh et al., 2010, Ching-Sung et al., 2007, Scanaill et al., 2006).

An example could be a smart system for remotely monitoring the patient's condition, which consists of a supervisory computer, a data router and a data logger (Kale et al., 2007). Once a patient's monitored physiological parameters (of a critical condition) breach predefined set limits, an SMS message and the mobile number(s) of the doctor(s) will be sent immediately by the server computer via the data router and the mobile phone. The doctor(s) can receive the critical messages within the coverage range of the GSM network.

Another GSM-based mobile telemedicine system has been developed for monitoring a patient's ECG and temperature signals (Tahat, 2009). In its client unit, the vital signals are acquired with limited sampling frequency, packed within a group of messages and then sent through an RS232 port by using a mobile phone. The sampling rate is restricted because of the specified SMS message limited size and quantity, which only allows discontinuous information and low precision signals. Besides, the consultation/server unit requires an advanced PDA and compatible application software to receive the signals. The block diagram of the client unit is excerpted and illustrated in Figure 2-10 (Tahat, 2009).

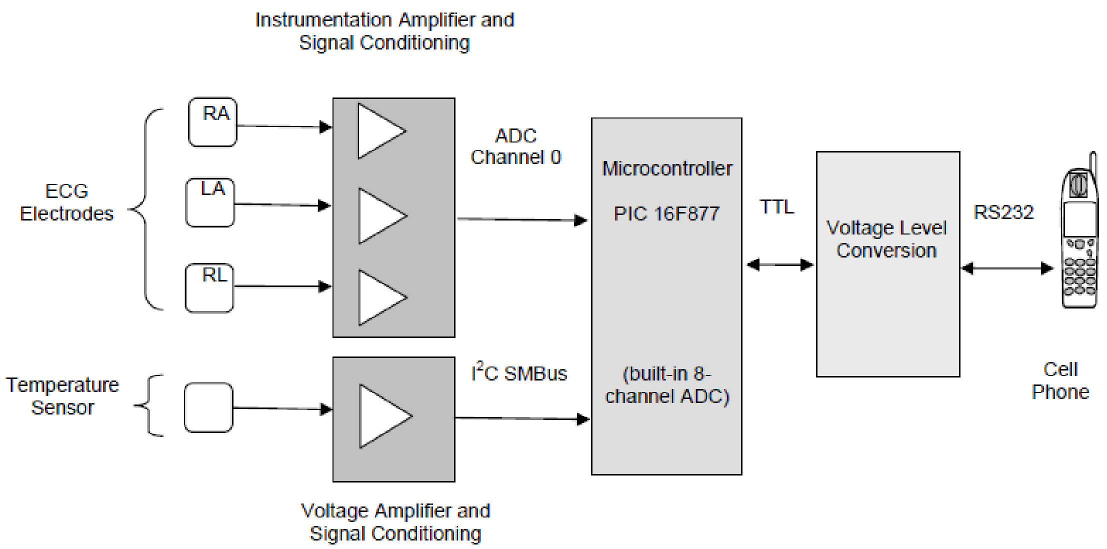


Figure 2-10 Block diagram of the client unit (Tahat, 2009)

An additional enhancement has been proposed by using a priority message layer added to the existing routing protocol in the system, which insures success of the messages delivery (Polk et al., 2007). Nevertheless, GSM network applications in telemedicine are severely constrained by its data transmission rate (up to 9.6 Kbps).

The general packet radio service (GPRS) adds a packet data service to GSM, supporting a theoretical maximum downlink data rate up to 171.2 Kbps. It is considered as so-called 2.5G. Compared to GSM, GPRS has a higher capacity and better mobile data services (Voskarides et al., 2003).

Rasid, M.F.A. and Woodward, B. describe a telemedicine processor system for multichannel biomedical signal transmission via GPRS networks (Rasid and Woodward, 2005). Signals including ECG, blood pressure, body temperature, and saturation of peripheral oxygen (SpO_2) are pre-processed by a signal conditioning module. The signals are then handled by a peripheral control module to be further processed and stored by a processor and memory module. Also the signals are fed to a Bluetooth communication module and are then transmitted to a mobile phone. Subsequently a GPRS network carries the signals to share them with clinicians who can connect to the GPRS cellular network. The prototype layout of the processor module is excerpted (Rasid and Woodward, 2005) in Figure 2-11.

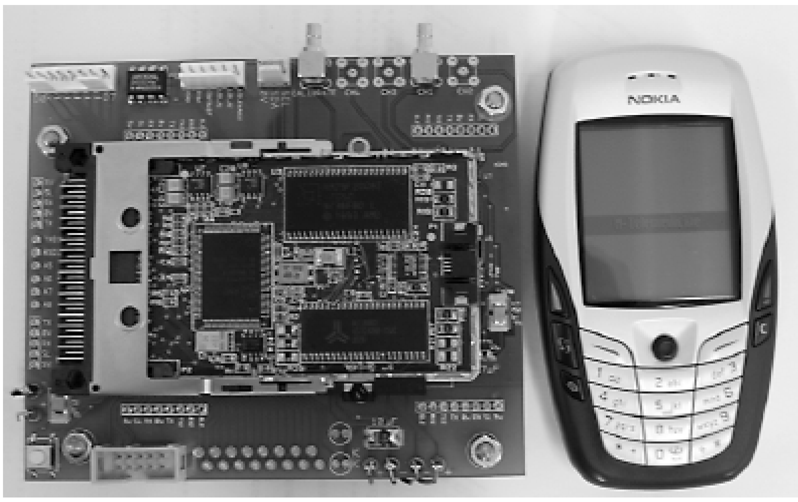


Figure 2-11   Prototype layout of processor module (Rasid and Woodward, 2005)

A new mobile ECG monitoring alert system was proposed by Dong-Her et al. in 2010. A signal acquisition device, radio frequency identification (RFID) reader and a GPRS module at the patient's side were used for sending alarm messages to a hospital server and database. Its system schematic is shown in Figure 2-12 (Dong-Her et al., 2010).

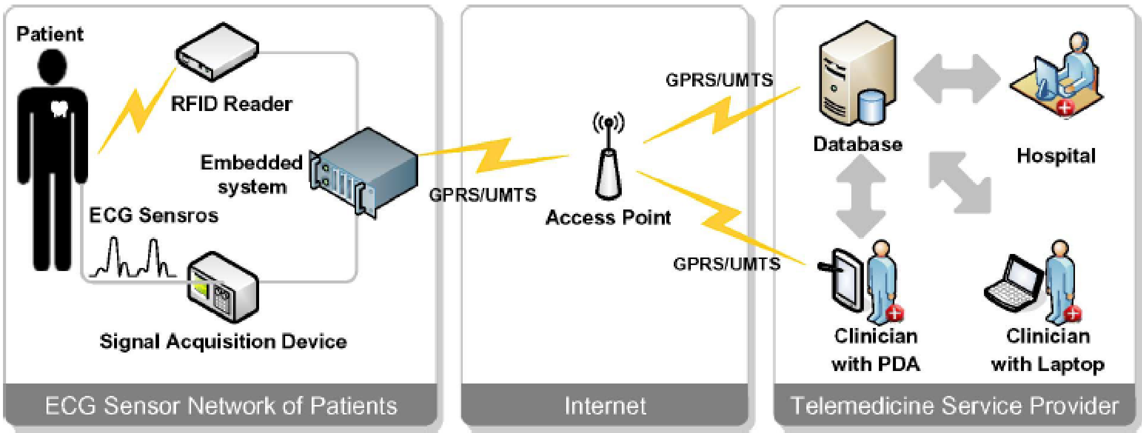


Figure 2-12   Schematic of a mobile ECG monitoring alert system (Dong-Her et al., 2010)

Again, the monitoring would not be continuous and real-time although a large volume of data can be sent in GPRS compared to GSM. The biomedical signals are required and transmitted only if expected by receipt of a command sent from the server.

In order to gain high speed data rate in wireless communications, 3G is designed to support wider bandwidth based on wideband CDMA. This new generation mobile communication standard is backward-compatible with existing mobile networks. By employing 3G, the transmission of broadband, real-time biomedical signals becomes possible over mobile wireless networks. Transmission rates of up to 10 Mbps can be achieved. The adoption of 3G in telemedicine brings massive enhancement respecting better performance, high mobility and flexibility. A new system model based on the 3G network has been introduced with the aims of providing effective mobile solutions (Liu et al., 2006, Fei et al., 2006). However, only preliminary designs and scenarios have been specified without any accomplished prototype at present.

For the support of roaming across heterogeneous wireless networks, such as cellular networks, WLAN, satellites links and so on, 4G wireless networks are being designed to communicate at very high bit rates in the range of 50 Mbps or more. Evolution of telemedicine systems via 4G networks is very promising and attractive, as 4G standards and service become reliable and pervasive.

Another promising wireless telecommunication technology is WiMAX which may be the main competitor to 4G as introduced in former section. However, WiMAX could be a complementary to, rather than a substitute for, 3G in such applications. With high data rate wireless connectivity, large-range mobility and guaranteed QoS, WiMAX technology could potentially proffer optimised solutions for telemedicine systems (Niyato et al., 2007). All emerging wireless networks in telemedicine share one disadvantage, that is, insufficiency of universal acceptance and ubiquitous deployment. Furthermore, nearly all endeavours derived from new generation mobile telecommunication services are still in the pilot stage.

Cellular networks were not originally designed for telemedicine systems but they can be immigrated for this purpose to provide mobile scenarios. Even though cellular networks offer a reasonable compromise between the mobility requirement and the cost of the system, there are still many issues which may influence the service quality. First of all, in the actuality of the cellular networks, coexistence of multiple standards (different generations) and utilisation of diverse access technologies (like TDMA or CDMA) have been one of the obstacles in sustaining improvement of cellular networks in telemedicine. Generally, cellular phones/smart phones/PDAs equipped with one access technology cannot be used to access other cellular networks with different ones. On the other hand, interoperability across multiple standards or generations is also a knotty problem as a result reining up the development in telemedicine by means of cellular networks. However, new generation mobile networks may facilitate the barrier but universal acceptance and deployment of their standards are still pending. Alternatively, designing compatible multi-adapter mobile devices will make the system complex and costly. Secondly, network coverage is still limited in less densely populated areas and network blind spots still exist. Therefore, it is evident that monitoring services may be less reliable in such areas.

Additionally, unstable network performance (such as error rates, handoff failures etc.) can directly affect the reliability of telemedicine systems. The peer-to-peer telecommunication mode curbs multi-access possibilities. Moreover, cellular networks use licensed frequencies so as that users have to assume subscriber expense (Niyato et al., 2007, Varshney, 2009).

2.4.3 The Internet in telemedicine

Modern telemedicine systems are associated with large amounts of biomedical information from patients, which is often transmitted over networks or mobile telephony systems. The rapid growth of wireless technologies and the Internet provides new opportunities to access medical services and biomedical information with enhanced mobility and efficiency. Similar to cellular networks, the Internet has been utilised as a subsequent transmission network linked to front-end short-range wireless connections in telemedicine systems.

The usage of the Internet aims at pervasive distribution of the telemedicine services, which stems from the fact that the Internet is the most global system of networks. The Internet is a worldwide, publicly accessible series of interconnected computer networks. As of June 30, 2012, the Internet usage statistics news and population statistics showed that there were over 2.4 billion users in the world (InternetWorldStats, 2012). The Internet is a network of networks, which means it can be regarded as a backbone network in architecture of telemedicine systems. By the utilisation of Ethernet, satisfactory transmission rates can be achieved for delivering physiological signals in real-time. The Internet as a communication medium offers flexible, easy-to-access and cost-effective remote systems which in turn have generated an increased demand in telemedicine applications.

Take one telemedicine system as an example: this wireless telemedicine system has a multi-parameter module which acquires a variety of signals including SpO_2, heart rate, temperature, blood pressure and ECG from the corresponding sensors. This group of data are serially transferred to a processing module and then fed to a Bluetooth module through

UART port. By using the Bluetooth protocol, a computer/laptop receives the data from the wireless module so that the patient's vital signs can be uploaded and stored to a server database in the monitoring centre. Hence patient's data can be accessed by doctors (Elgharably et al., 2008). Other similar design makes use of PDAs to obtain the biomedical signals and transmit data to the Internet (Zhi and Guanglie, 2007). A new telemedicine system for the home monitoring of lung function was designed based on a real-time signal processing algorithm and Internet-based network architecture (Dellaca et al., 2009). The telemedicine device consists of a main board which converts the signals to digital form; it also controls peripherals, as well as a data processing. The transmission board processes, records and transmits the signals in real-time. Using the Internet (TCP/IP standard) or GPRS, the data are finally received and administered by a central data server. A sketch map of the network architecture is excerpted (Dellaca et al., 2009) in Figure 2-13.

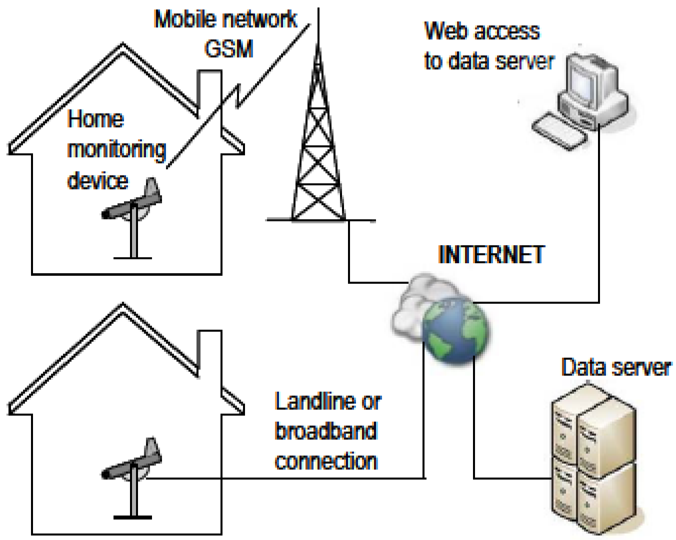


Figure 2-13 Network architecture of the telemedicine system for lung function (Dellaca et al., 2009)

Recently, a proposed remote monitoring system designed by Dilmaghani et al. (2011) advances the research further step. A node of the wireless sensor network (wireless patient portable unit) and an access point (wireless access point unit) were created, which eliminates the need for a PC as a transmission gateway on the patient side. Through the Internet, ECG signals can be received by an SQL database server and an ECG Windows server in the hospital. The system scenario is illustrated in Figure 2-14 (Dilmaghani et al., 2011).

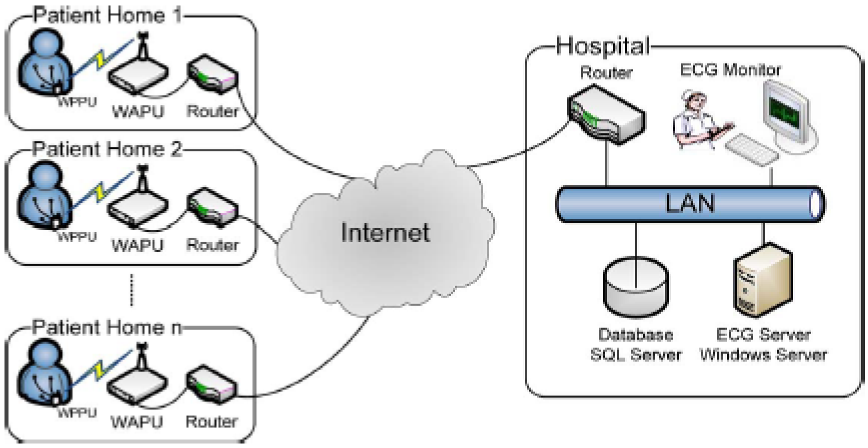


Figure 2-14 Network architecture of the remote monitoring system for ECG (Dilmaghani et al., 2011)

To date, a host of Internet enabled telemedicine systems are evolving from pilot proposals to prototypes. The upsurge of interest in real-time monitoring, Internet accessibility and wireless solution expedites the development of telemedicine systems (Pantelopoulos and Bourbakis, 2010, Min et al., 2011). A range of representative telemedicine systems with Internet connectivity have been evaluated and are summarised in the following table, based on their individual features and architectures.

Table 2-2 A list of representative telemedicine systems with the Internet connectivity

Description of the system	Front-end module	Transmission gateway	Monitoring centre	Applied data representation
Intelligent home care ECG system (D'Angelo et al., 2010)	Commercial ECG recorder	Local PC	Jetty web server	Off-line ECG

Fetal ECG monitoring system (Ibrahimy et al., 2011)	A commercial ECG front-end module ‘CARDIC’	Local PC	Remote PC	Off-line fetal ECG
Portable Multi-life-parameter monitoring system (Qinwu et al., 2011)	Physiological signal collection unit	PC or GPRS module	Server and database	Real-time ECG, body temperature, blood pressure and blood sugar
CARA healthcare system (Bingchuan and Herbert, 2011)	Wireless monitoring device	PC	CARA server and database	Real-time ECG, temperature and SpO_2
A Zigbee-based telecardiology system (Ernest et al., 2011)	CSN801 multi-parameter module	PC and Zigbee module	Hospital server	Real-time ECG
Pervasive cardiac monitoring system (Hai-ying and Kun-mean, 2010)	Wireless ECG sensor	Local access server	Remote access and surveillance server	Real-time 4-lead ECG
A secure and resource-aware BSN system (Honggang et al., 2010)	ECG device	PDA	Server machine	Real-time ECG
A wireless monitoring system (Rotariu et al., 2011)	A wireless body area network of devices	PDA	Central medical database server	Real-time alarms for ECG, SpO_2, body temperature, respiration, blood pressure

Wireless sensor networks (Dilmaghani et al., 2011)	Wireless patient portable unit	Wireless -access point unit	Windows server and database SQL server in the hospital	Real-time ECG

Almost all Internet-based telemedicine systems are based on a common architecture, i.e.: the front-end signal acquisition and conditioning module that detects the biomedical signals; a short-range wired or wireless connection to the gateway or the local control terminal on the patient side; the transmission gateway that delivers the data via the Internet to the database server in the hospital or the monitoring centre; the database server stores and manages the patient's data. In this way, doctors or clinicians can visit the database server to access the patient's data remotely through the Internet. It can be seen from recent work that many such systems are still in their infancy. What is more, most researches have not slipped the leash of applying a host PC/laptop/smart phone/PDA or other bulky devices as transmission gateways on the patient side to the backbone network, i.e., the Internet. This constrains the patient's mobility and the device's portability. In addition, the usage of a central database server in the hospital or the monitoring centre also inflates the system complexity and the requirement for maintenance.

2.5 Summary

In this chapter, comprehensive up-to-date research in the application of telemedicine has been presented and analysed. In the section which discussed DSP technologies in relation to the various clinical applications, the major biomedical applications for ECG were elaborated on. To extract the biomedical signals from dedicated sensors, signal processing algorithms are employed and implemented by the front-end signal conditioning module. As the technology advances, the trend is moving from portable device such as the Holter monitor, sophisticated (and costly) implantable technology, towards non-invasive, comfortable and wearable intelligent garments. Wireless networking systems play the increasingly prominent role in modern telemedicine systems. The short-range wireless technologies are employed to transmit signals acquired by the front-end module and forward them to the backbone networks (the cellular networks or the Internet). The transmission gateways are the retransfer adapters between them. However, the current telemedicine systems would have required bulky equipment/computer/laptop/smart phone/PDA as the gateway, which undoubtedly effaces many advantages of telemedicine systems. This led to the concept of an integrated portable telemedicine system within a single platform to address the drawbacks discussed above and offer a promising solution to these limitations. That is what this research focuses on.

CHAPTER 3

Theory

3.1 Introduction

Considering the design scenario of the Internet-based real-time DSP system for telemedicine, this chapter discusses fundamental principles in the various fields which support the development of the system. The discussion is centred on the system as a tool for remote biomedical signal processing. It involves principles and analysis of biomedical signals, algorithms and theories of signal processing methods as well as protocols and mechanism of network transmission models. These interpretations in the following sections cover the major and principal parts of the system.

3.2 Principles and analysis of ECG

The ECG is a transthoracic interpretation of the electrical activity of the heart over time, which is externally captured and noninvasively recorded by skin-attached electrodes and associated instrumentation. It is essential for diagnosing cardiopathy that indicates the overall rhythm of the heart in order to find weaknesses in different parts of the heart muscle (Varshney, 2009). This section will introduce how heart contraction operates and forms the ECG signal. The arrangement of ECG leads and the morphology of ECG waveforms are also described here.

3.2.1 Operating mechanism of heart contraction

The heart has four chambers — right atrium, left atrium, right ventricle and left ventricle. During atrial or ventricular diastole of the heart, blood is coming into any of them, which refers to re-polarisation. During ventricular systole of the heart, a sequence of contractions forces the blood from the heart to the aorta. The Figure 3-1 depicts a normal human heart (Khan, 2008, Varshney, 2009).

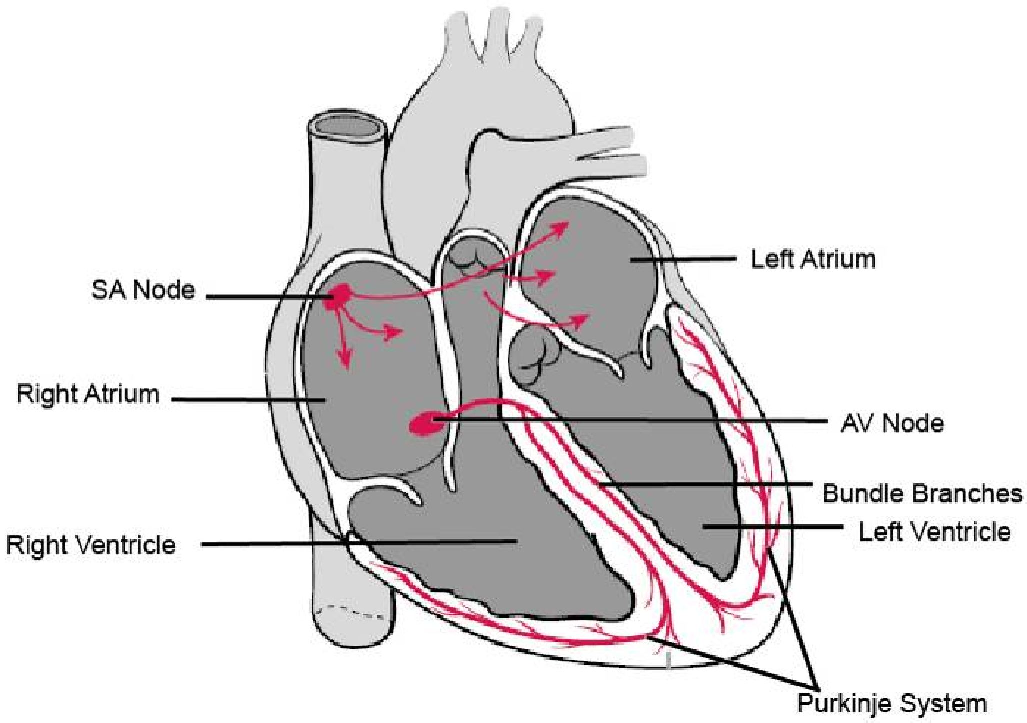


Figure 3-1 Conformation of normal heart (Khan, 2008)

To ascertain the mechanism of each contraction of the heart, the genesis of the electrical activity in the myocardial cells needs to be interpreted. The excitatory waves of electrical activity, which precede every contraction of the heart, stem from the cross-membrane movement of ions between the intracellular and extracellular compartment of the myocardial cells. The myocardial cells provide energy to maintain ionic concentration differences between the inner and outer surface of the cell membrane. Normally, Na^+ and

Ca^{2+} are more concentrated in the extracellular compartment but a higher concentration of K^+ appears in the intracellular compartment. The cell membrane has the functionality of selective channels which conduct diffusion of Na^+, Ca^{2+} or K^+ ions systematically, which are triggered principally by specific voltage changes (Goldberger. and Ng., 2010, Khan, 2008).

The cardiac muscle cells have two states: resting and action potential. In the resting cardiac muscle cells, the channels for K^+ open and K^+ ions efflux from the cells. Compared to the extracellular compartment, the intercellular compartment of the cells is electrically negative with a potential difference of about -90 mV. In the state of action potential, the channels for Na^+ and Ca^{2+} open so as that the concentration of these two ions decreases in the extracellular membrane and increase in the inner one. That is to say, Na^+ and Ca^{2+} diffuse into the cells so that the intracellular compartment is positive in contrast to the extracellular compartment. In the following figure 3-2, corresponding to a curve of the varied voltage, the progress of the electrical activity in the myocardial cells is illustrated (Khan, 2008).

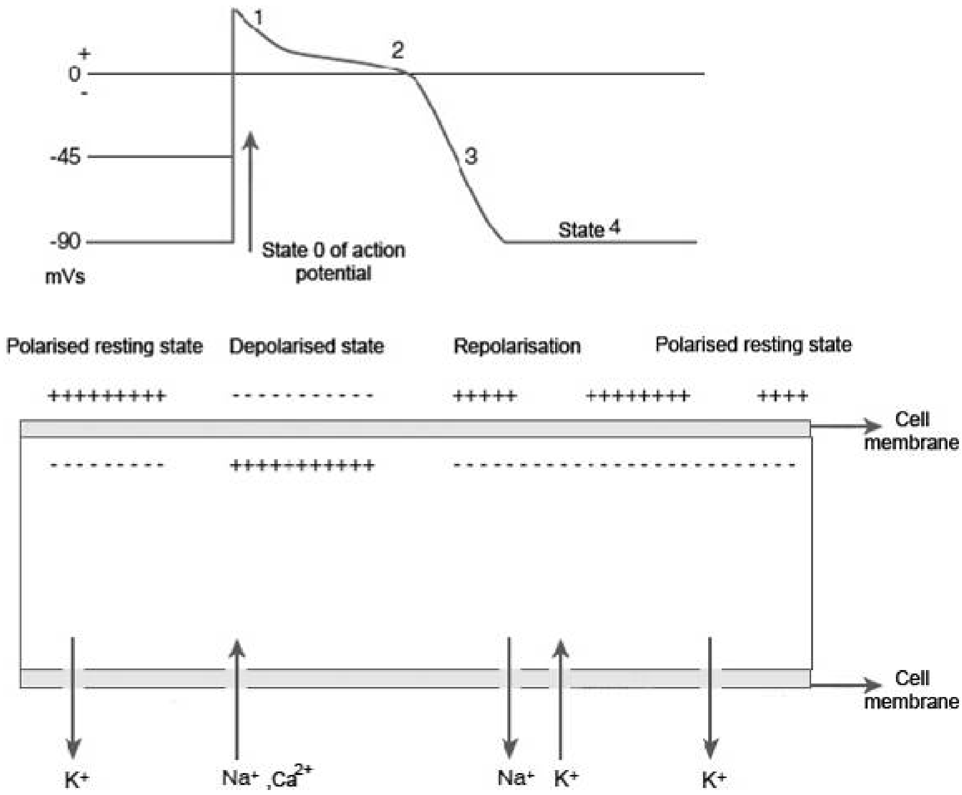


Figure 3-2 Progress of the electrical activity in the myocardial cells (Khan, 2008)

The figure depicts clearly the procession of the electrical activity of the myocardial cells. The myocardial cell in the resting state is so-called polarised. While the excitatory waves stimulate the cells, massive positive ions like Na^+ and Ca^{2+} move into the cells – it is step 0 of action potential. The following step 1, 2 are referred to as depolarisation. In the stage of repolarisation (step 3 of action potential), the sodium-potassium ATPase pump transfers Na^+ inward and K^+ outward. Afterwards, the membrane is in electrical balance and the cells recover back to the resting state (Khan, 2008).

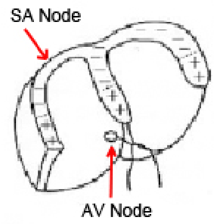
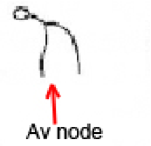
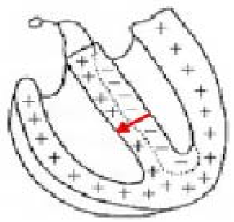
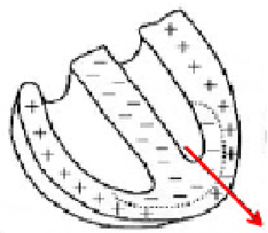
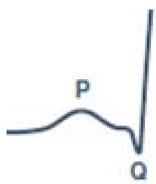
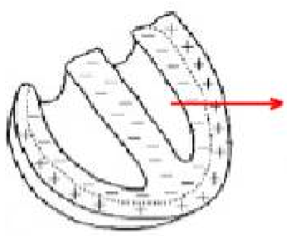
A sequence of voltage changes are engendered between the inner and outer surface of the cell membranes, as the selective channels of the membranes open and close in a sequential and cyclical manner. The action potential transports Ca^{2+} into the cardiac muscle cells, which triggers contraction of the heart. The propagation of the action potential from one myocardial cell to the others results in a sequence of contraction from one part of the heart to the other. It traverses the heart and initiates the systole.

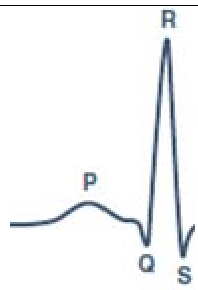
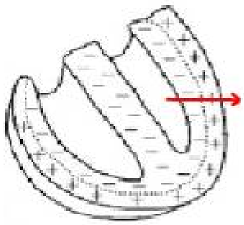
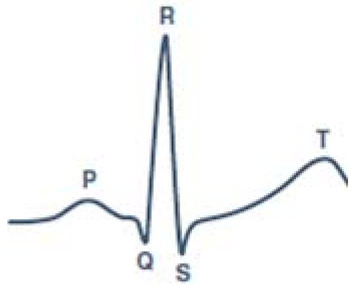
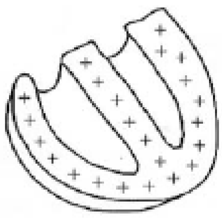
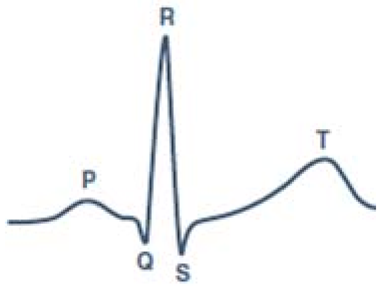
3.2.2 Formation mechanism of ECG signal

As explained in the last section, each cardiac muscle cell has a sequence of voltage changes across the cell membrane. The process of this potential difference towards zero is identified as depolarisation. It activates the mechanism in the cardiac muscle cells which evokes the heart to contract.

In the period of each contraction, portions of a healthy heart have a systematic progression of a wave of depolarisation. This sequence of action potential needs to be triggered by an electrical impulse from a pacemaker with automatic, spontaneous and cyclical functionality. The sinoatrial (SA) node is predominant pacemaker providing the electrical impulse, which has this property of automaticity. It is a group of cells on the wall of the right atrium. Normally, the impulse emanates from the SA node and passes via an intrinsic conduction pathway within the heart. It spreads out through the right and left atrium towards the atrioventricular (AV) node. The propagation of the impulse is physiologically delayed in the AV node. It is necessary because it allows a transient time for blood flowing from the contracting atrium to the ventricles before the systole. It also reduces the rapid SA node beat rate to slower ventricular beat rate. Thence, rapid conduction continues into the ventricles along the right and left bundle branches, into the Purkinje system, and then the entire ventricular myocardium (the bundle branches and the Purkinje system are displayed in Figure 3.1). It ultimately induces the initiation of the ventricular systole (Khan, 2008, Goldberger. and Ng., 2010, Vecht et al., 2009). The table below elaborates on the proceeding of the propagation of the electrical activity (marked as red arrows) as well as the corresponding portions of the ECG signal (Rosendorff et al., 2006, Varshney, 2009, Iaizzo et al., 2009, Khan, 2008).

Table 3-1 Propagation of electrical activity and counterpoint of ECG signal (Iaizzo et al., 2009)

Proceeding description	Sectional view of heart	Portions of ECG
Depolarisation in atrium		
Transient halt at AV node		
Depolarisation in septum		
Depolarisation along bundle branches		
Depolarisation along the Purkinje system		

Depolarised ventricles		
Repolarisation in ventricles		
Repolarised ventricles		

This excitatory wave propagation of electrical activity is detected as tiny rises and falls in the voltage - in the level of mV, between two electrodes set either side of the heart. The ECG signals are acquired principally by means of detecting the feeble voltage changes on the body skin that are caused while the depolarisation happens during each heartbeat or contraction.

3.2.3 ECG leads

Two electrodes acting as a differential pair set either side of the heart on the surface of the skin, which are termed an ECG lead and yield one channel of ECG signal. Section 2.2 generally introduced delineation of the 12 ECG leads. Ten electrodes which form the 12 channels and their placement were also enumerated. They are labelled: RA, LA, RL, LL,

V_1, V_2, V_3, V_4, V_5 and V_6. The twelve ECG leads are sorted in terms of different electrode pairs or configurations as three groups: standard limb leads, augmented limb leads and precordial leads. The standard limb leads include Lead I, II and III. The augmented limb leads consist of Lead aVR, aVL and aVF. Lead V_1 – V_6 make up of the precordial leads (Vecht et al., 2009, Khan, 2008).

The twelve ECG leads represent the electrical activity of the heart on two planes. The three standard limb leads and the three augmented limb leads portray the electrical activity on the frontal plane. The six precordial leads depict it on the horizontal plane (Foster, 2007). Therefore a three-dimensional perspective to the heart is established to monitor the heart. The illustration in Figure 3-3, known as Einthoven's triangle, clearly indicates the arrangement of the electrodes' position and the leads' placement.

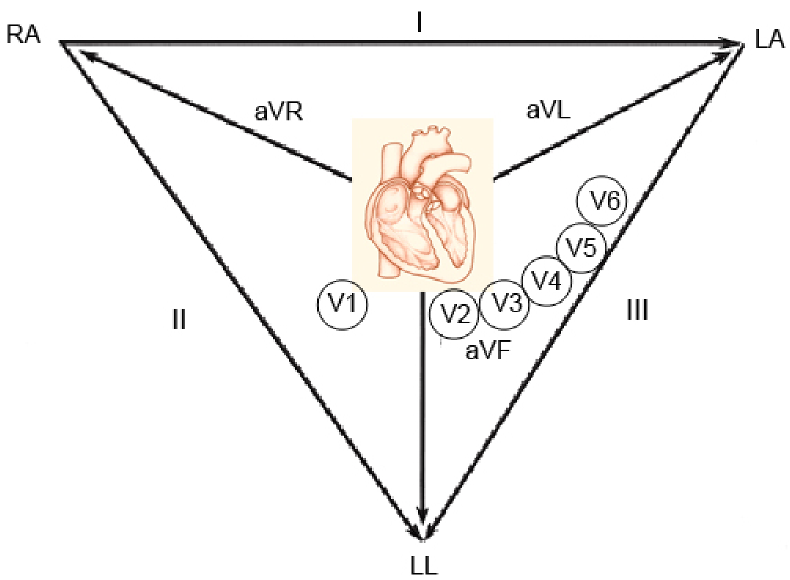


Figure 3-3 The Einthoven's triangle

In the illustration, the heart is considered to be located at the centre of the triangle. The electrode RA, LA and LL form the vertices of the Einthoven's triangle. They represent both the standard limb leads and the augmented ones. The equations for each lead are stated as follows (Iaizzo et al., 2009).

As the standard limb leads employ a pair of electrodes on either side of the heart, Lead I, II and III refer to the bipolar leads. Lead I can be expressed as the voltage difference between the electrode LA and RA:

$$\textbf{\textit{Lead I = LA – RA}} \tag{3-1}$$

Lead II is the voltage difference between the electrode LL and RA:

$$\textbf{\textit{Lead II = LL – RA}} \tag{3-2}$$

Lead III is the voltage difference between the electrode LL and LA:

$$\textbf{\textit{Lead III = LL – LA}} \tag{3-3}$$

The augmented limb leads view the heart activity from different vectors but also stemming from the same three electrodes LA, LL and RA. Each of the augmented limb leads is composed of one limb electrode and a neutral reference lead which is combined with the other two limb electrodes. Therefore, Lead aVR, aVL and aVF are unipolar leads. Lead aVR connects one electrode on RA and another electrode on a combination of LA and LL. It can be presented as:

$$\textit{\textbf{Lead aVR}} = \textit{\textbf{RA}} - \tfrac{1}{2}\,\textit{\textbf{(LA + LL)}} \tag{3-4}$$

Lead aVL connects one electrode on LA and another electrode on a combination of RA and LL, which states as:

$$\textit{\textbf{Lead aVL}} = \textit{\textbf{LA}} - \tfrac{1}{2}\,\textit{\textbf{(RA + LL)}} \tag{3-5}$$

Lead aVF connects one electrode on LL and another electrode on a combination of RA and LA. The expression can be:

$$\textit{\textbf{Lead aVF}} = \textit{\textbf{LL}} - \tfrac{1}{2}\,\textit{\textbf{(RA + LA)}} \tag{3-6}$$

The precordial leads are placed directly on the front of the left thorax to detect the electrical activity of the heart on the horizontal plane. They are also considered as unipolar leads where Wilson's central terminal is used as a negative electrode which is the combination of the electrode RA, LL and LA. Wilson's central terminal W_V calculation equation is:

$$\mathbf{W_V = \tfrac{1}{3}\,(RA + LA + LL)} \qquad (3\text{-}7)$$

Three bipolar standard limb leads, three unipolar augmented limb leads and six unipolar precordial leads comprise the full twelve ECG leads, which provide the evaluation of the cardiac electrical activity through the heart with the purpose of diagnoses. In the applications of monitoring, only Lead I or/and Lead II is/are required.

3.2.4 Analysis of the ECG waveform

The key segments of the ECG complex are: P-wave, PR interval, QRS complex, ST segment, T-wave, etc. These are depicted in Figure 3-4 (Iaizzo et al., 2009, Khan, 2008).

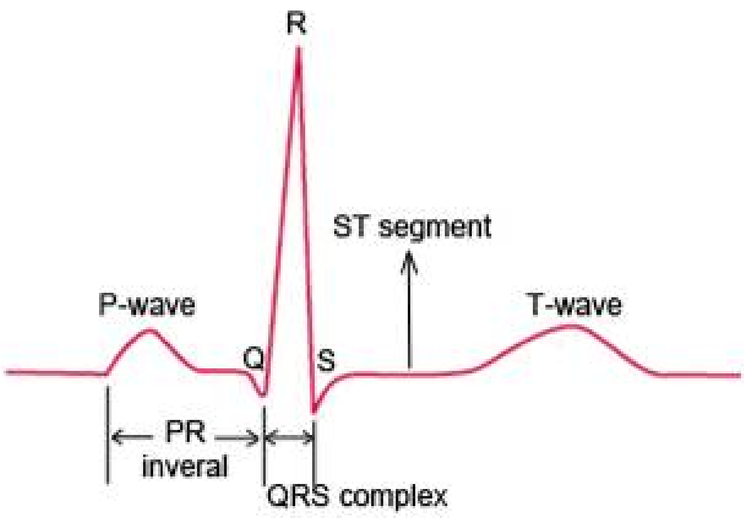


Figure 3-4 Clinical interests in ECG waveform

The P-wave represents the electrical activity occurring in the right and left atria which are stimulated from the SA node. The PR interval is defined as the time when the electrical impulse conducts from the atrium, to the AV node, across the ventricular septum, through the bundle branches, into the Purkinje system until the beginning of the ventricles' depolarisation. The QRS complex is a large spiky deflection of the ECG signal, which interprets the most rapidly changing portions of the signal. It represents the depolarisation proceeding of the ventricular myocardium. Specifically, it represents the high-frequency components of the ECG signal. After the QRS complex and before the T-wave, the ST segment manifests as an imperceptibly ascending curve which dictates the depolarised stage of the ventricles. At the end of the sequence, the T-wave represents the repolarisation of the ventricles. The whole curve portrays the waveform of a standard ECG signal (Khan, 2008, Iaizzo et al., 2009).

Most of the diagnostic information is contained below 100 Hz for adults and below 250 Hz for infants (Iaizzo et al., 2009). The frequency spectrum of the ECG signals needs to be determined in order to retain the clinical information by noise removal and signal extraction. The relevant discussion refers to section 4.3.2 'Multiple filtering'.

3.3 Elements of ECG signal detection

The ECG signal is the most vital sign of the patient's cardiovascular status. The origin of the ECG signal is ionic current flow of nervous stimuli and muscle contractions, which generate bio-potentials that are conducted onto the body surface. To measure the electric potential on the surface of thoracic tissue, proprietary biopotential electrodes and leads are employed.

3.3.1 Ag/AgCl electrodes vs. dry electrodes for ECG signals

Typical ECG electrodes are based on the common Ag/AgCl design (which incorporate electrolytic gel-adhesive), or new dry belt-type or textile electrodes.

Several issues surrounding electrodes have been discussed in section 2.3.3, the most crucial one of which is excessive movement artefact. As the biological signals measured by the electrodes are the superposition of all biopotential activities from the body, motion artefacts produced by muscle movements are considered as interference and need to be suppressed. Dry electrodes also move across the skin, resulting in much more artefacts than those associated with gel-adhesive electrodes due to impedance changes. Movement artefacts associated with dry electrodes bring considerable interference resulting in the need for complicated signal processing to obtain acceptable interpretation of ECG signals (Searle and Kirkup, 2000). In the initial period of an electrode’s operating life, Ag/AgCl electrodes with electrolytic gel-adhesive present much lower movement artefact levels than dry electrodes.

Electrode contact impedance is another unavoidable issue that is often addressed in clinical applications. Firstly, each ECG lead is composed of a pair of electrodes contacting the skin and two wires linked to the output device. It is notable that the skin conductivity is different over the body. Consequently, contact impedance mismatch between the electrode pair exists in practice. According to Ohm’s law, the contact impedance mismatch on the two leads results in an unwanted differential voltage on the amplifier input. Secondly, as known, low source impedance and high load impedance ensures the transfer of higher electrical voltage (but with limited power); this is termed impedance bridging. By this means, signal degradation during transmission from electrode to signal conditioning module can be effectively minimised. Therefore, for high-fidelity measurement and amplification of the bio-potentials, the input impedance (or load impedance) of the amplifier must be much greater than the source impedance from the electrodes (Geddes and Baker, 1966, Webster, 1998). Hence good electrode contact is essential. This results in a low source impedance and, when combined with a high input impedance amplifier, ensures signal integrity.

In brief, excessive movement artefact and electrode contact impedance are the most severe obstacles facing the new dry electrode technologies. The stability and quality of ECG signals obtained using Ag/AgCl electrodes at present exceeds those produced using dry electrodes, at least during the initial stage of their operating life (Searle and Kirkup, 2000). Moreover, the deployment of dry electrodes results in escalating complexity of the increased overheads associated with the signal processing. Therefore in this design the Ag/AgCl electrode with electrolytic gel-adhesive is employed to sense the ECG signals rather than the new dry electrode. A picture of the front and rear face of a typical Ag/AgCl electrode is shown in Figure 3-5, one of which connects to the skin and another links the lead connected to the signal conditioning module.

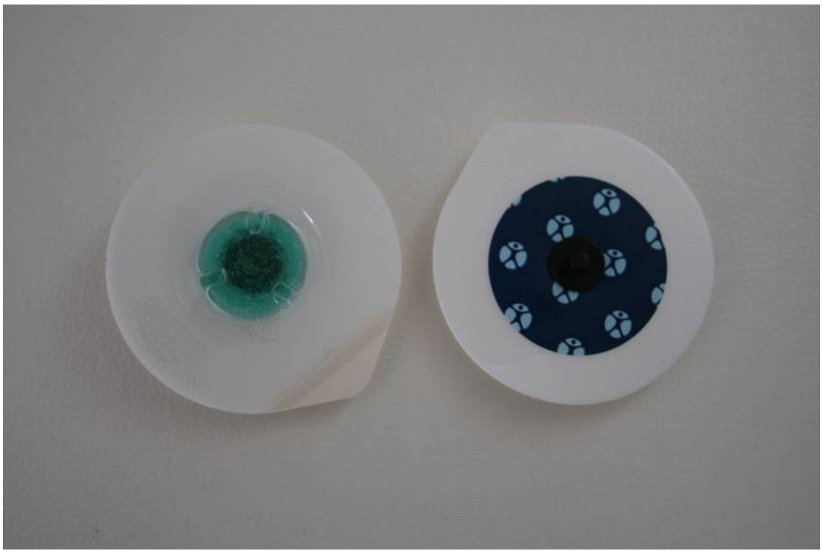


Figure 3-5 Two sides of the Ag/AgCl electrode with electrolytic gel-adhesive

This is also known as metal-plate type disposable electrode. The diameter of the Ag/AgCl electrode pad is 50 mm and its thickness is about 2 mm. On one side of the electrode pad, in the pad's centre, is a silver-plated disk covered with an AgCl layer which serves as the electrode. Electrolytic gel is coated on the disk with adhesive foam in order to maintain close contact with the skin. One of the important tasks of the electrolytic gel is to reduce the variability of the contact impedance. On the other side of the pad, a silver-plated snap attaches to the disk, which connects to the lead cable. Since the electrolyte gel slowly dries, it must be noted that the performance of the Ag/AgCl electrodes degrades over time. Thus, measurements should be taken during the recommended operating window of the electrode.

Nonetheless, it should be stated that the deployment of the innovative dry electrodes is an irrevocable tendency in the application of telemedicine for the prolonged monitoring of biomedical signals. If the new generation of dry sensors/electrodes reaches the performance of the Ag/AgCl electrodes, this designed system can effortlessly employ the dry electrodes.

3.3.2 ECG lead cables

The main task of ECG lead cables is to conduct the weak raw ECG signals to the front-end signal conditioning circuit. To assemble with the electrode, the female portion of the lead cable is snapped onto the male end of the electrode. ECG leads apply pairs of the electrodes to measure the potential differentials. The picture is a view of the ECG lead cables in Figure 3-6.

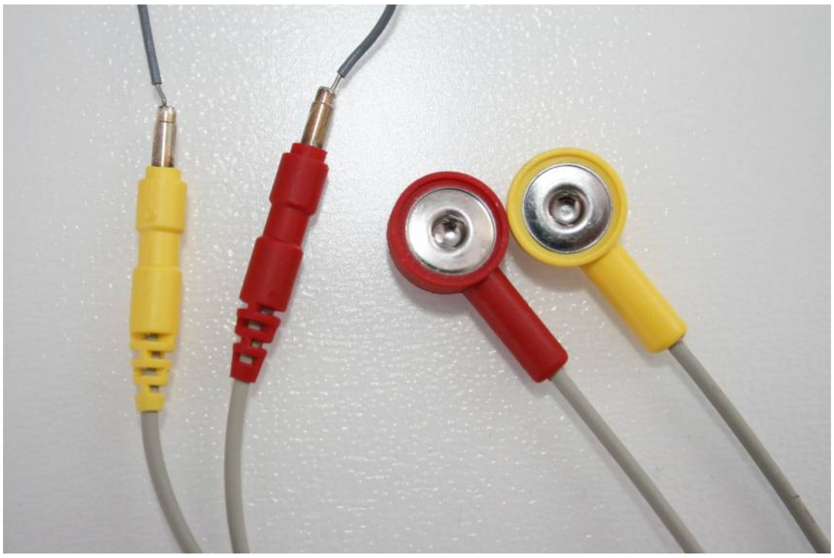


Figure 3-6 A view of the dedicated ECG lead cables

It is important that two electrodes in the same pair should be identical in order to produce faithful signal, as the same to two ECG lead cables in the same pair. If dissimilar material is used in contact, interference arising from the half-cell potential difference effect may be evoked, inducing electrode contact impedance mismatch. Consequently, detrimental noise will be increased.

Excessive movement of the cable can result in noise or even electrode removal. The lead cables ought to be attached flexibly but securely. To achieve this, the lead cable can be taped to the skin, leaving a loose section between the electrode and the taped point in order to relieve the strain.

Since the ECG lead consists of a pair of cables, magnetic induction interference from power lines can be induced in the signals by the coupling of electric and magnetic fields. To eliminate such interference, twisted pairs may be used. That is, two cables are wound around each other along their lengths. Additional shielding over the twisted pair will also provide better suppression of electromagnetic radiation noise (Reynders et al., 2005, Webster, 1998).

3.4 Signal processing theory

After the ECG signals are detected by the electrodes and conducted via the leads, a series of signal processing operations to restore the biomedical signals is not optional but vital. This section will discuss the relationship between signal and system, followed by considerations of amplification, how Fourier analysis impacts signal processing, filtering techniques and elements of analogue to digital conversion.

3.4.1 Signal and system

For better understanding of how the biomedical signals are manipulated in the designed system, it is necessary to introduce qualitative explanations of signal processing and linear system principles.

A signal is a set of information or data which varies with time, space or other variable(s). Signals may be classed in various ways. In this case, untreated biomedical signals are detected as analogue signals which vary with time. As the independent variable is continuous time t, the signals can be denoted as function $x(t)$. After conditioning and processing, ultimately the signals will be converted into digital format for further interpretation and storage.

Here, a system is defined as an entity which manipulates a set of signals to yield another set of signals; the system can be either physical hardware or conceptual software, even both. In this work, the input signals are the untreated ECG signals detected by the electrodes and the output signals are the digital ones which are received and stored at the client-end over the Internet. The function block of the signal and system is portrayed in Figure 3-7.

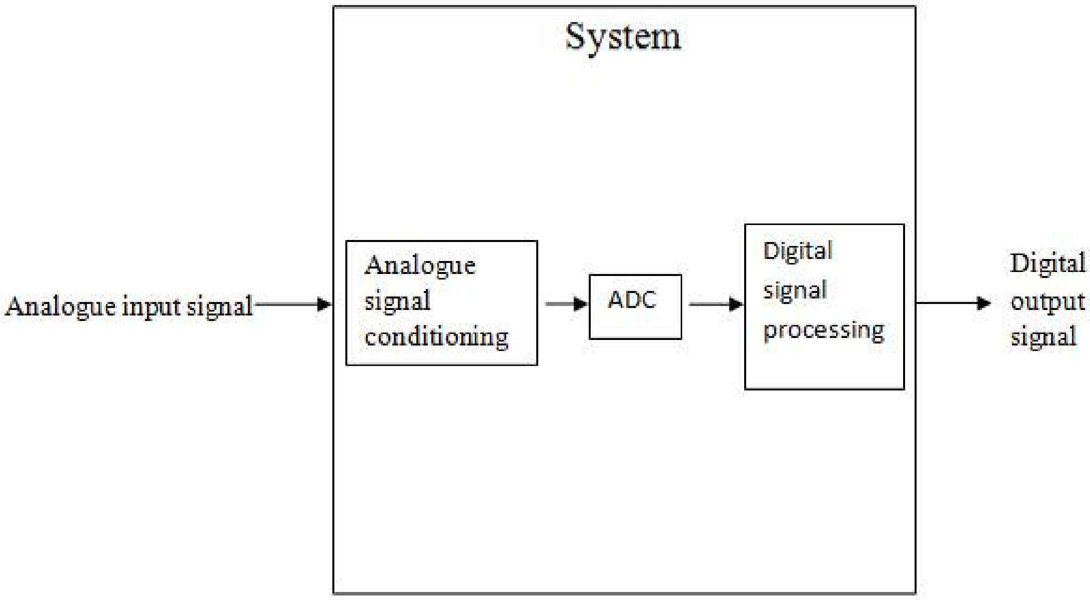


Figure 3-7 Function block of the signal and system

This system has characteristics of temporal and spatial invariance as well as uniqueness, which means that it always products a unique output in response to a unique input. The behaviour of the system shows superposition since it can support multiple channels.

3.4.2 CMRR and amplification

The common-mode rejection ratio has significance in amplification. As the raw ECG signals have features of small voltage range, large voltage offset and large common-mode components, a high CMRR is important during signal amplification. An instrumentation amplifier with a high CMRR can amplify the ECG signals and also suppress common-mode noise — one of the main sources of interference.

The CMRR depicts the tendency of output voltage to change while both differential inputs are changed by equal amounts. If an instrumentation amplifier has two differential inputs V_+ and V_-, and yields an output V_o. Then the output V_o can be given as (Webster, 1998):

$$V_o = G_d\left(V_+ - V_-\right) \tag{3-8}$$

Where G_d is the differential gain. In practise, the output of the instrumentation amplifier is interpreted as:

$$V_o = G_d\left(V_+ - V_-\right) + \frac{1}{2}G_{cm}\left(V_+ + V_-\right) \tag{3-9}$$

Where G_{cm} is the common-mode gain. Thus, the CMRR can be defined as the ratio of the powers of the differential gain over the common-mode gain:

$$CMRR = 20\lg\left(\frac{G_d}{\left|G_{cm}\right|}\right) \tag{3-10}$$

In general, high differential gain and low common-mode gain are desired to attain a high value of the CMRR, which signifies that less common-mode noise will appear at the output. The CMRR has normally frequency-depended specification regarding the input signals.

The instrumentation amplifier is designed with a high CMRR value in order to reduce the effect of common-mode noise. The Analog Devices AD620 instrumentation amplifier was used in this work. The gain of the AD620 is set using a single external resistor R_G. The calculation equation of the gain G can be expressed as (Analog Devices, 2004):

$$G = \frac{49.4k\Omega}{R_G} + 1 \tag{3-11}$$

For a given gain, R_G is determined by (Analog Devices, 2004):

$$R_G = \frac{49.4k\Omega}{G-1} \tag{3-12}$$

The theoretical value of R_G for amplifying gain of 1000 is 49.45 Ω according to the equation (3-12). In practice, a resistor of 47.5 Ω yields a gain of 1041 by equation (3-11).

3.4.3 Fourier analysis

Signals are normally present in temporal domain as waveforms by first observation. In order to investigate signals' frequency spectrum, Fourier analysis and synthesis undoubtedly lie at the nucleus of the theorem in signal processing. Fourier analysis explicates that any periodic waveform in time domain may be decomposed into a sum of harmonically sinusoids and/or cosinusoids, each of which is of particular magnitude and phase. Signals may be periodic or aperiodic, continuous or discrete. According to this, the Fourier analysis technique may be categorised by four varieties: the continuous Fourier series; the continuous Fourier transform; the discrete Fourier transform; the discrete time Fourier transform (Gaydecki, 2004). Since the ECG signals can be considered as periodic, and signals in both analogue and digital forms are operated in this work. The continuous Fourier series and the discrete Fourier transform will be concerned with the following discussion.

The continuous Fourier series handles a periodic, continuous signal in time domain and yields a series of aperiodic discrete Fourier harmonics in frequency domain. That is to say, a continuous periodic waveform can be interpreted as a series of sine and/or cosine factors. Its synthesis equation can be denoted in trigonometric form as:

$$x(t) = A_0 + \sum_{n=1}^{\infty} B_n \cos\frac{2\pi nt}{T} + \sum_{n=1}^{\infty} C_n \sin\frac{2\pi nt}{T} \tag{3-13}$$

Where T is the period of the waveform; the coefficient A_0 represents the DC level of the signal; the coefficient B_n represents the magnitude of the cosine factor of the nth harmonic; the coefficient C_n represents the magnitude of the sine factor of the nth harmonic. The fundamental harmonic may be written as $\omega_0 = \frac{2\pi}{T}$.

The synthesis equation of the continuous Fourier series may be defined in the exponential manifestation:

$$x(t) = \sum_{n=-\infty}^{\infty} X[n] e^{jn\omega_0 t} \tag{3-14}$$

The coefficient $X[n]$ can be depicted as the complex exponential analysis equation:

$$X[n] = \frac{1}{T}\int_0^T x(t) e^{-jn\omega_0 t} dt \tag{3-15}$$

The discrete Fourier transform (DFT) handles a periodic, discrete signal in time domain and yields a series of periodic discrete Fourier harmonics in frequency domain. By the same way as stated above, the DFT synthesis equation in trigonometric form is expressed as:

$$x[k] = A_0 + \sum_{n=1}^{N/2} X_r[n]\cos\left(\frac{2\pi kn}{N}\right) + X_i[n]\sin\left(\frac{2\pi kn}{N}\right) \tag{3-16}$$

Likewise, the DFT synthesis equation or inverse DFT (IDFT) may be expressed in the exponential equivalent:

$$x[k] = \sum_{n=0}^{N-1} X[n] e^{j2\pi kn/N} \tag{3-17}$$

And DFT exponential analysis equation:

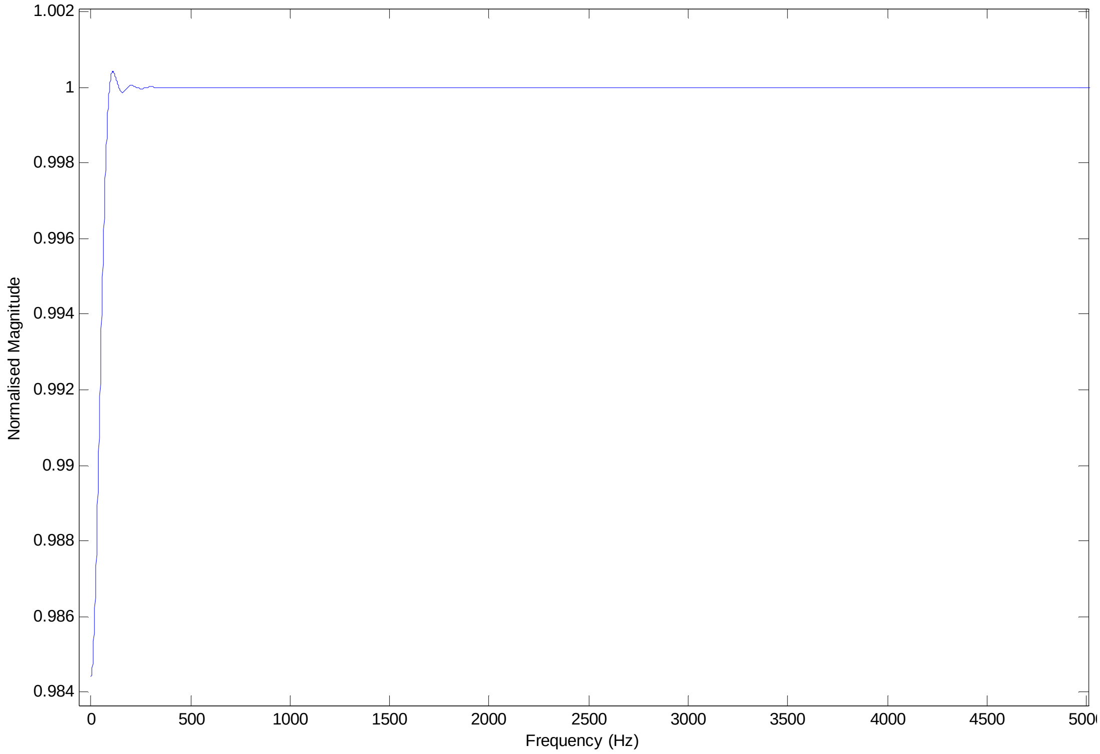
$$X[n] = \frac{1}{N}\sum_{k=0}^{N-1} x[k] e^{j2\pi kn/N} \tag{3-18}$$

As the frequency spectrum of the signal is evaluated, filtering the redundant frequency components of the signal can be executed by a variety of filters. Fourier analysis and synthesis merely map the signal into a different domain and do not alter the information content.

3.4.4 Methods in signal filtering

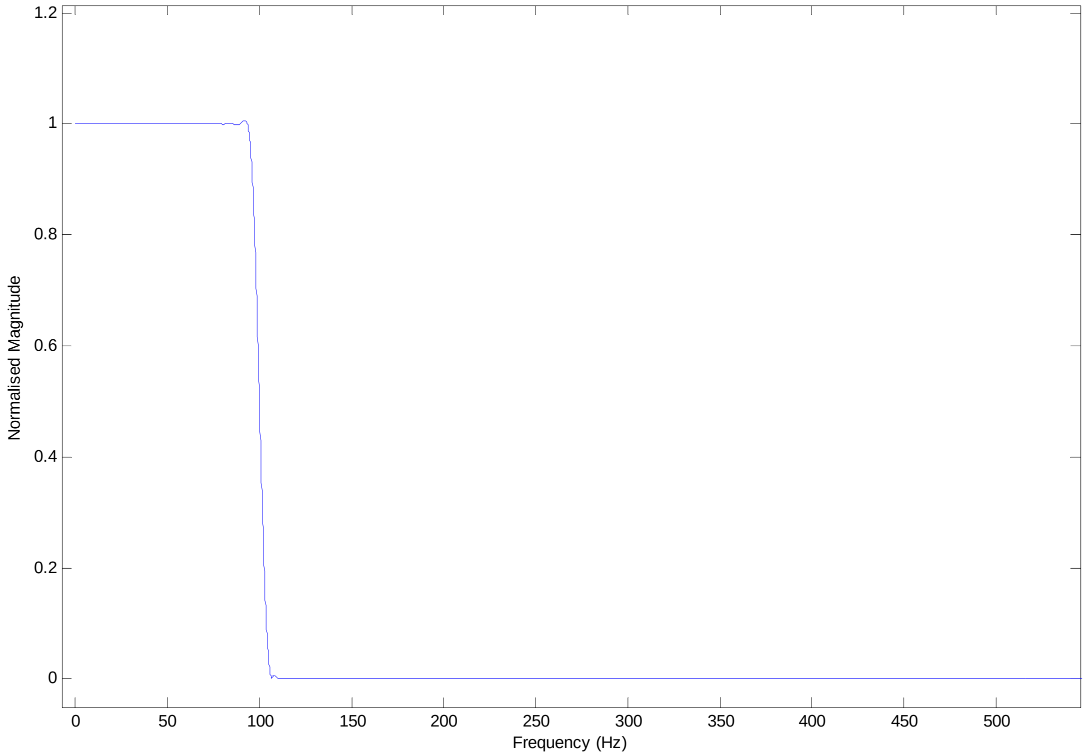
Signal processing is primarily utilised to analyse, modify and extract information from signals, for instance, to remove interference or noise from signals. In the context of signal processing, signal filtering is a process in which a device or processing system suppresses redundant frequency components from signals, and passes or augments required ones. Signal filtering may be performed using analogue or digital methods. In the designed system, both are involved to manipulate the signals in different forms.

Frequency selective filters are designed to pass signals and reject noise, based on desired frequency components. These filters can be classified on the strength of the selected frequency bands. Although there are many, the ones applied here are low-pass, high-pass and notch filters. The frequency responses of these filters were simulated by the Signal Wizard software package using a high-pass filter with a cut-off frequency of 0.5 Hz, a low-pass filter with a cut-off frequency of 100 Hz and a notch filter set at 50 Hz respectively. The representative illustrations of these filters' frequency responses are shown in Figure 3-8.



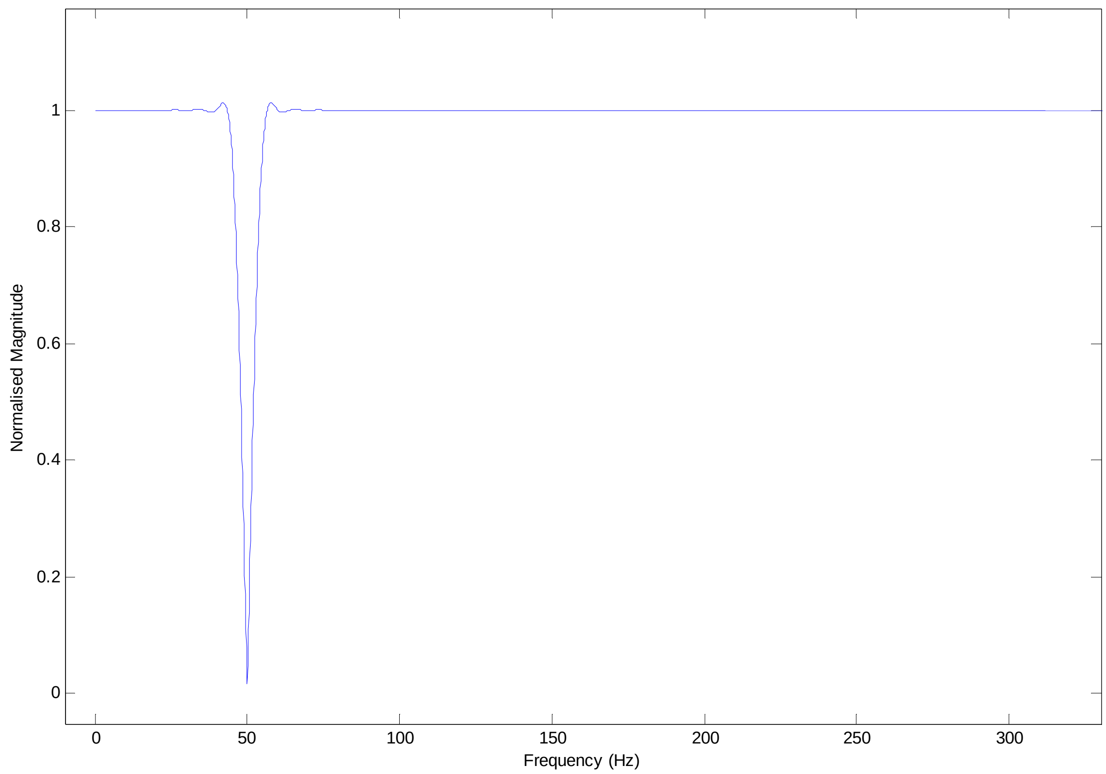
Normalised Magnitude
1.002
1
0.998
0.996
0.994
0.992
0.99
0.988
0.986
0.984
0
500
1000
1500
2000
2500
3000
3500
4000
4500
Frequency (Hz)

(a)



Normalised Magnitude
1.2
1
0.8
0.6
0.4
0.2
0
0
50
100
150
200
250
300
350
400
450
500
Frequency (Hz)

(b)



(c)

Figure 3-8 Frequency responses of the representative (a) high-pass filter, (b) low-pass filter, and (c) notch filter

The high-pass filter attenuates frequency components less than the cut-off frequency and passes ones higher than it. Typically, the cut-off frequency is measured point where the attenuation is greater than 3 dB, namely, the amplitude of the output reaches $1/\sqrt{2}$ of the input's amplitude. The low-pass filter attenuates frequency harmonics beyond the cut-off frequency and passes ones lower than it. The notch filter just rejects a selected frequency band, passing all the others.

As the biomedical signals processed by the analogue filters are still corrupted by random noise to some extent, the signals represent a slightly spiked curve. The Savitzky-Golay digital filter is employed to smooth the curve based on a polynomial of a given order. The algorithm of the Savitzky-Golay smoothing filter for digital signals is to replace each value of a series of data with a new value yielded by a polynomial fit. The new value is a local average of surrounding data points. By this means, the filter can reduce the level of noise with limited influence over the original value of the data. The Equation 3-19 denotes the signal smoothing algorithm of the Savitzky-Golay filter (Bromba and Ziegler, 1981):

$$\left(x_K\right)_N = \frac{\sum_{i=-n}^{n} C_i x_{K+1}}{\sum_{i=-n}^{n} C_i} \tag{3-19}$$

Where x_K is original signal, $\left(x_K\right)_N$ is a smoothed data point, C_i is polynomial coefficient, and n is the filter width. When each smoothed data yield, there are $2n+1$ neighbouring points numerically handled. The signal-to-noise ratio (SNR) of the signals can be improved by the implementation of the Savitzky-Golay digital smoothing filter (Luo et al., 2005).

3.4.5 Elements of analogue to digital conversion

Analogue to digital conversion is necessary to convert signals from analogue to digital form. The central steps of the conversion involve sampling, quantisation and encoding. Firstly, the analogue signal is sampled and converted into a discrete-time but continuous-amplitude format. Secondly, the sampled signal is quantised by the level of 2^n, where n is number of bits of the ADC. Now the signal is discrete both in terms of amplitude and time. Subsequently, the quantised samples are encoded into distinct binary words, i.e., the universally accepted digital signals. The following function block in Figure 3-9 shows the process.

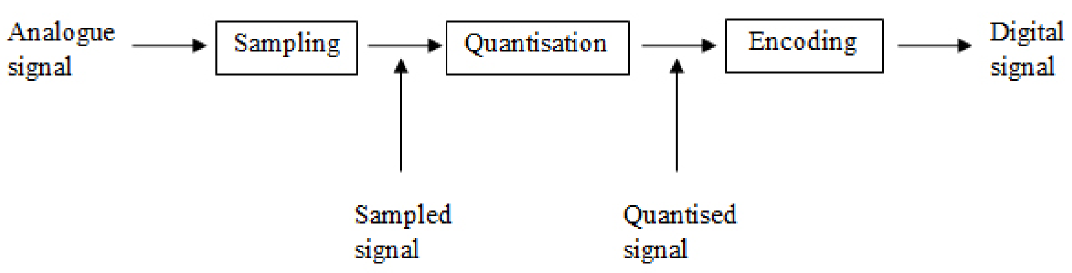


Figure 3-9 Function block of analogue to digital conversion

Sampling can be described as a process in which an analogue signal (continuous in both time and amplitude) is acquired at discrete-time intervals. If viewed in equation form, the sampling process may be written as:

$$x_{discrete}(t) = x_{analogue}(t)\sum_{n=0}^{\infty}\delta(t - nT) \qquad (3\text{-}20)$$

Where $x_{discrete}(t)$ denotes the sampled signal, $x_{analogue}(t)$ is the band limited analogue signal, $\delta(t - nT)$ is a series of shifted impulse functions, and T represents the sampling interval occurring at n =0, 1, 2….

The Nyquist theorem or the Shannon sampling theorem stipulates that, if the highest frequency component of an analogue signal is named f_0, the signal must be sampled at the rate of at least $2f_0$ to interpret the signal with completed fidelity. Failure to sample at a rate of twice the desired frequency leads to signal aliasing. Nevertheless, it is common for the input signal to incorporate frequency harmonics beyond the frequency scope of interest; such components can be considered as noise. If these noise components are included, the sampled signal will be contaminated or degraded. For that reason, analogue anti-aliasing filtering is applied before sampling in practice (Ifeachor and Jervis, 2001, Gaydecki, 2004).

3.5 Communication network protocols and mechanism

A suite of communication protocols support the network transmission of the digital biomedical signals onto the Internet. A communication protocol is a set of standard rules for data representation, signal transmission, authentication and error detection required to transmit information over a communications channel. Each employed network (e.g. the Ethernet, the Internet or Wi-Fi in this context) or each application layer has its corresponding protocol, for example, TCP/IP, IEEE802.11 or Hyper-Text Transport Protocol (HTTP). This section will elaborate on these protocols and their operating mechanisms.

3.5.1 Client-server model

A client-server module denotes a pattern of network communication which organises shared information or files from servers to distributed clients. The model is adopted in this system. Each instance of the client software sends data requests to the connected server. The server can accept these requests, process them, and then return the requested information or files back to the client. The advantage of the client-server architecture enables the information on the server to be distributed among multiple independent users. In this case, the stand-alone hardware with its firmware performs the role of the server. The clients can be any web browser or homologous software accessible by the Internet.

3.5.2 The OSI model

The seven-layer Open Systems Interconnection (OSI) reference model is an abstract description of network functions published by the International Organisation for Standardisation (ISO). The OSI model formulates architecture of the standard protocols, which aims to provide complete interoperability among networks. It divides network communications into seven layers: Application, Presentation, Session, Transport, Network, Data link and Physical, sequentially from top to bottom. Each layer performs functions which are necessary to accomplish the error-free exchange of data between network users. Figure 3-10 illustrates the arrangement of the OSI model seven layers.

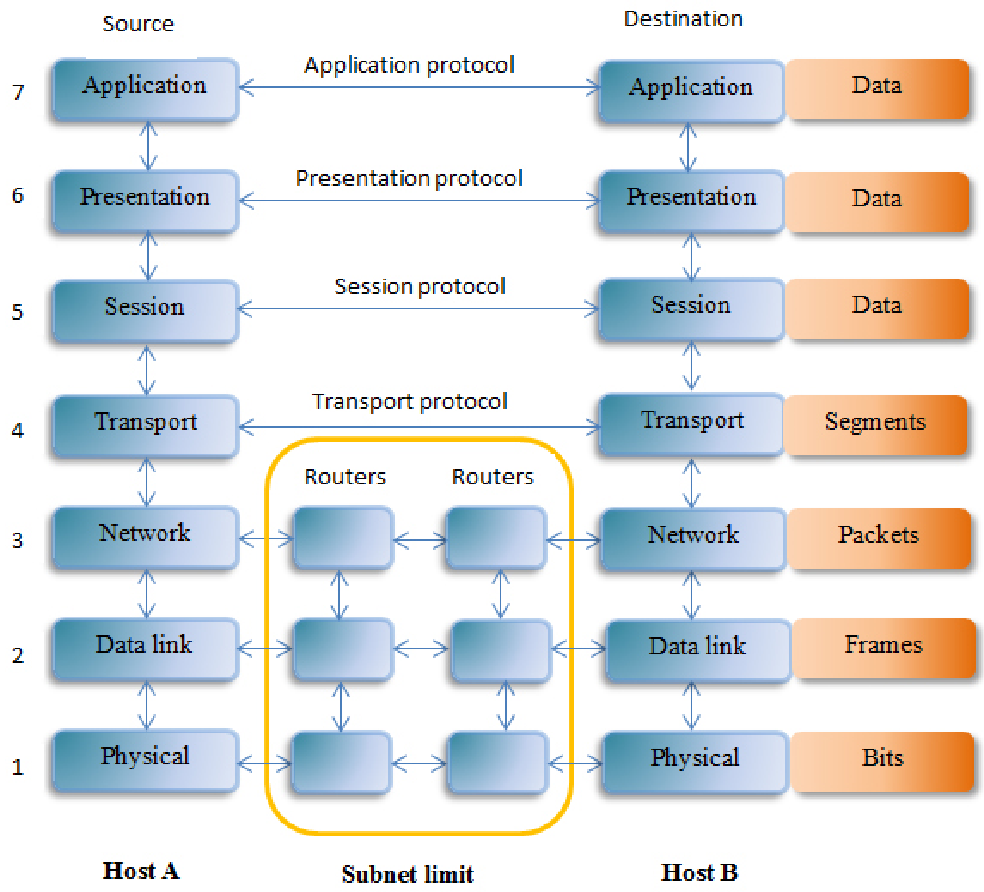


Figure 3-10 The OSI model seven layers

• The Application layer

The highest layer in the OSI model is Layer 7 – the Application layer. It is the closest layer for the end user to access the network service, which interacts with the user by using a software application interface. This layer offers definitions of several types of applications. For example, the identity and availability of the communication partner for data transmission may be determined by the application layer. It can determine the availability of the resource network or requested communication and also manage synchronising the communication. The application layer contains a variety of common and custom protocols, such as the HTTP and the File Transfer Protocol (FTP).

• The Presentation layer

This layer makes it possible for communication with different transmitted data representations, acting like a ‘data translator’. It is devoted to transform data from a network form into the acceptable form toward the upper layer, i.e., the application layer. The service of this layer defines data formatting, data compression as well as data encryption. The data are represented in variable syntaxes or different character codes, so that they need to be formatted according to this layer’s service. This layer’s function also provides data compression and encryption algorithms to improve the data transmission rate and the data security.

• The Session layer

The session layer establishes, manages and terminates the dialogues/sessions between local and remote entities. A session refers to a process of two entities’ applications to establish, maintain and terminate a connection. The services in this layer define how the messages will be transferred: half-duplex or full-duplex. If a large amount of massages are to be sent, function of synchronisation points supplied by the layer can partition the sequence of messages into groups. The session layer also performs security functions like user recognition and logging.

• The Transport layer

The services of this layer provide transparent transfer of data between users. The transport layer offers flow control, segmentation/reassembly and error control for reliable sequenced data delivery. Within this layer, the transfer rate of data is controlled and adjusted during network buffers. If the session layer passes a large size message onto the transport layer, the message will be decomposed into a packet size which the network layer accepts. In the transport layer of the destination entity, those packets belonging to the same message are reassembled in sequence. The acknowledgement of the successful data transmission is provided in this layer. Typical protocols in the Transport layer may be Transmission Control Protocol (TCP) and User Datagram Protocol (UDP).

• The Network layer

The network layer is concerned with addressing and multiplexing which splits a single communications channel into multiple time divided communications channels. It formats the data into the packets with a header containing the packet sequence and the destination address. This layer mainly performs routing functions responsible for the routing of the packets across a network. Based on the layer, a router can send data throughout the extended network such as the Internet. The representative protocol regarding to the network layer is Internet Protocol (IP).

• The Data Link layer

This layer provides functions to arrange the input data bits from the lower layer into data frames (i.e., logical sequences), to transmit the frames sequentially between multiple network nodes, and perhaps to detect errors occurring in the physical layer. In the frame, a header which contains source and destination addresses indicates the originator and recipient. If the errors are detected, the service of the layer requests retransmitting the unacknowledged frames from the source node. The Data Link layer supports both Wide Area Network (WAN) and LAN. In detail, the Data Link layer may be depicted as two sub-layers: the Logical Link Control (LLC) sub-layer and the Media Access Control (MAC) sub-layer. This layer is the foundation of the dominant Ethernet architecture — the IEEE 802.3 specification. It will be introduced in the latter section.

• The Physical layer

The bottom OSI layer is the physical layer, which defines the electrical and physical connections and the means of transmitting the binary data between nodes over a communications channel. It describes the interactions between a single device and a physical medium, including voltage levels of the binary 0s and 1s, connectors and pins assignments, cable specifications and so on. The primary services which this layer provides are: establishment and termination of a connection to a transmission medium; modulation or encoding between the representation of the binary data pattern and the corresponding signals propagated in the physical cabling or the radio link; regulation of connectors and pins assignments for various medium specifications (Muller, 2003, Walrand and Varaiya, 2000, Cowley, 2007).

The architecture of the OSI model represents and maps network functions or services in a concise manner. Notwithstanding, the OSI model is not practical network architecture because it does not specify the specific services and protocols to be operated in each layer. Many networks do not strictly compel compliance with the OSI model so as that it is rarely used nowadays. The model just tells what each layer is supposed to perform as a guide or a reference. The widely used abstraction of protocols and services may be based on the TCP/IP model.

3.5.3 TCP/IP model

The Transmission Control Protocol and the Internet Protocol suite or model is the set of communication protocols used for networks, especially, the Internet. The TCP/IP model characterises functionalities of the software application, the end-to-end transport connection, the internetworking range and the scope of the direct links to other nodes. This model provides service of the end-to-end Internet connectivity which stipulates how data should be formatted, directed, shipped, routed and delivered to the right destination. It specifies various protocols for different types of communication between entities and provides a framework for more detailed standards.

The TCP/IP model is composed of a set of layers: the Application layer, the Transport layer, the Internet layer and the Network Access layer, sequentially from top to bottom (Tanenbaum, 2002). Each layer resolves several problems of transmitting data and offers a well-defined service to the upper layer protocols.

The TCP/IP model is often comparable to the OSI model, whereas the design concept of the TCP/IP model is not of rigidly conformity of the OSI reference layering scenario. The

Application layer in the TCP/IP model performs the combined functions of the Application layer, the Presentation layer and most of the Session layer in the OSI model. The TCP/IP model's Transport layer involves the close service of the OSI model's Session and Transport layers. The Internet layer is much alike a sub-layer of the OSI Network layer. The Network Access layer underneath performs as a counterpart of the OSI Data Link layer, Physical layer as well as parts of the Network layer (Cowley, 2007). The side-by-side comparison of the layers in two models is shown in Figure 3-11.

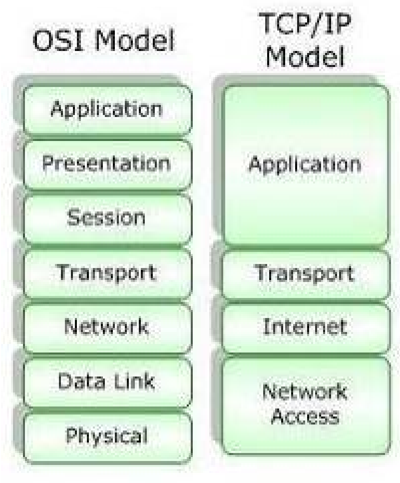


Figure 3-11 The OSI model seven layers and the TCP/IP model four layers

• The Application layer

In this layer, user data are created and coded according to application layer protocols, and subsequently are encapsulated into the Transport layer. The Application layer normally uses the lower level layers protocols to fulfil actual data transfer or to provide a stable network connection. Upon the layer, higher level protocols like HTTP are operated to communicate between peers (network communication hosts).

• The Transport layer

The Transport layer defines a transport mechanism, allowing peers on the source and destination hosts to build up an end-to-end conversation. Through the utilisation of 'ports', the Transport layer provides services of opening and maintaining connections between the Internet hosts, flow control, error correction as well as application addressing. Transmitting applications of this layer may be implemented as either TCP or UDP. In the context of the Transmission Control Protocol, it addresses several responsibilities to offers reliable byte stream, including orderly data delivery, correctness, discarding duplicate data, resenting discarded packets and traffic congestion control.

• The Internet layer

The task of the Internet layer is to exchange datagram across the source network to the destination entity. It defines the addressing and routing structures within the TCP/IP model, or in another words, it establishes the Internet. The principal protocol in this layer is the Internet Protocol, which performs functions of routing datagrams or data packets from the source network to the destination network. According to the Internet Protocol, host IP addressing and identification are also accomplished.

• The Network Access layer

This lowest component layer in the model describes the local network topology and the interfaces of two different entities. The Network Access layer is used to transmit and receive packets on the same local network link without routers. This layer performs functions of adding a packet header and then transmitting the frame over a physical connection.

The TCP/IP model is also known the Internet Protocol suite since the set of communications protocols is applied especially for the Internet. Hardware and software implementation based on the different protocols upon the layers may be diverse. The TCP/IP model is neither principal design criterion nor design reference for the operation of the Internet. It merely originates a framework of the logical groups and scopes of functions in the design of the Internetworking protocols.

3.5.4 HTTP

In the framework of the Internet Protocol suite, HTTP is a communication protocol used to transfer web pages over the Internet as the Application layer protocol. It is concerned with the interface to the TCP/IP stack upon the lower layers.

HTTP defines the communications that occur between the server and the client. The HTTP client accomplishes task of receiving and reading web pages and the HTTP server is used for storing and transferring them. HTTP relies on the transport protocol — TCP in this case, to support and guarantee data transmission from the server to the client, since it does not provide acknowledgement or retry service.

HTTP has a request-response mechanism. Generally, the client sends an HTTP request message to request a web page (or a file) by using the GET method. It normally employs a web browser or equivalent software to obtain the web page. The server which stores the file resource replies with an HTTP header followed by a message containing the file contents. The sketch in Figure 3-12 illustrates the mechanism of HTTP transmission between the client and the server.

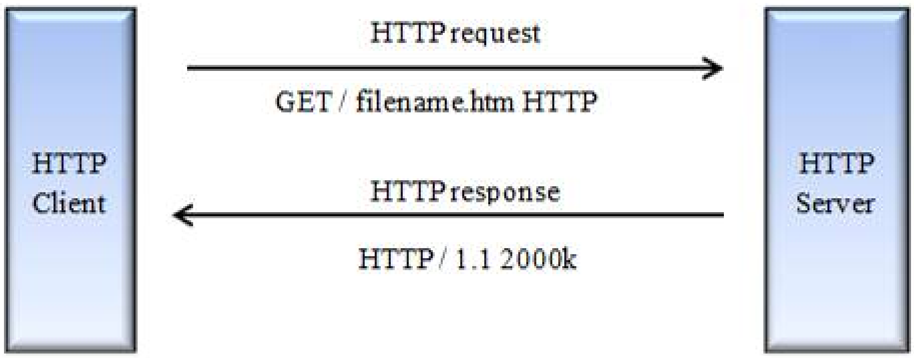


Figure 3-12   HTTP Request/Response mechanism

The HTTP server may require a single thread, which handles multiple sessions in the way of executing one session per pass. An HTTP session refers to a sequence of communication request-response transactions. The HTTP server establishes and listens for a connection request on port 80. When it receives a connection, the port is removed from port 80 to a random port so that port 80 can again await other connections. After that, a new HTTP session is created in the session array.

3.5.5 The Ethernet

The Ethernet is a predominant computer networking technology for LANs. The Ethernet, standardized by IEEE 802.3 specification, is defined by the physical layer and the MAC sub-layer of wired Local Area Network. It specifies a mechanical or electrical connection between devices and a multi-node addressable communications protocol (Walrand and Varaiya, 2000). This allows for robust data transmission across relatively long distances. The Ethernet is normally used with high-level protocols such as TCP/IP.

A common addressing format and a variety of MAC procedures are designated by the Ethernet. In the circumstances of the Ethernet LANs, the network interface of each user station is assigned a unique 48-bit MAC address which is contained in the data packet. The address specifies the source and the destination so that the packet can be dispatched to the appointed network interface of the station rather than the others. Regardless of how the Ethernet standards evolve, the same frame formats of all Ethernet generations remains in order to be interconnected. The frame format in IEEE 802.3 specification is described in details as follows.

Table 3-2 The IEEE 802.3 frame format

7	1	6	6	2	46-1500	4
Preamble	*Start Frame Delimiter*	*Destination Address*	*Source Address*	*Length/ Type*	*LLC Header & Data & Pad*	*Frame Check Sequence (CRC)*

At the beginning of the frame format, 7-bit Preamble dictates timing information and single bit Start Frame Delimiter tells the end of the timing bits. Destination address and Source address locate the sending and receiving station. The following two sections of the frame incorporate any headers for other protocols and preload data. The frame ends with a cyclic redundancy check (CRC) named Frame Check Sequence which detects the validity of the data (Cowley, 2007).

Moreover, a number of wiring and signalling standards are dictated by the Ethernet upon the Physical layer. The common standards are 10BASE-T and 100BASE-TX where the stream transmitting rates are 10Mbit/s and 100Mbit/s respectively. The physical media interfaces of both types make use of twisted pair cables and 8 position 8 contact (8P8C) module connectors or often termed RJ45. The illustration of RJ45 socket is shown in Figure 3-13.

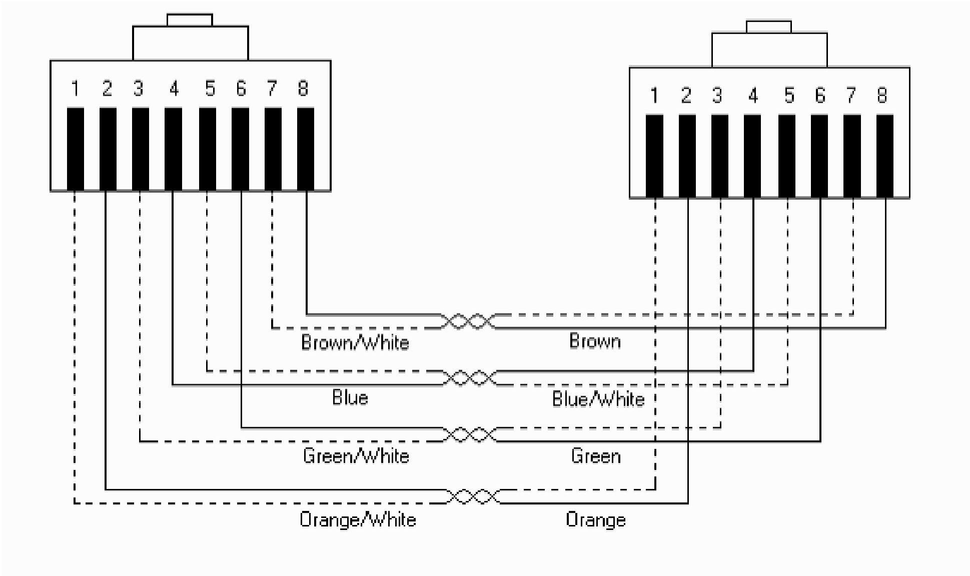


Figure 3-13 RJ45 socket connector

The Ethernet can also provide not only LANs over twisted pair cables but also WANs over fiber optic variants, as it is of compatibility with the WAN Physical layer standard (Cowley, 2007).

3.5.6 WLAN, IEEE 802.11 and Wi-Fi

A Wireless Local Area Network provides a connection of multiple entities by means of wireless distribution method, usually linking the Internet via an access point. It offers users the mobility and connectivity within coverage scope to the network, which becomes the prevailing option of the 'last mile' for the entire Internet infrastructure.

The IEEE 802.11 specification, which is a suite of protocols to implement WLAN communication, defines the services on the physical layer and the MAC layer. IEEE 802.11 is of two operating modes: Ad hoc mode supports peer-to-peer communication between two devices; infrastructure mode is the more common WLAN applications where the users communicate via an access point to a wired network infrastructure, such as the Internet. The IEEE 802.11 family constitutes a series of technique standards, in which the

IEEE 802.11b, g and n are prevalent. The following table may elaborate on the specifications (Cowley, 2007, Varshney, 2009).

Table 3-3 The specifications of IEEE 802.11b, g and n

<table>
<tr><th>802.11 standard</th><th>Operating Frequency (GHz)</th><th>Bandwidth (MHz)</th><th>Maximum data rate (Mbit/s)</th><th>Approximate indoor range (M)</th><th>Approximate outdoor range (M)</th></tr>
<tr><td>b</td><td>2.4</td><td>20</td><td>11</td><td>38</td><td>140</td></tr>
<tr><td>g</td><td>2.4</td><td>20</td><td>54</td><td>38</td><td>140</td></tr>
<tr><td rowspan="2">n</td><td>2.4</td><td>20</td><td>72.2</td><td rowspan="2">70</td><td rowspan="2">250</td></tr>
<tr><td>5</td><td>40</td><td>150</td></tr>
</table>

Wi-Fi is originally neither a technical term nor a protocol, but a trademark of the Wi-Fi Alliance. However, based on the IEEE 802.11 specification Wi-Fi has been widely accepted to depict a number of WLAN connectivity technologies (Labiod et al., 2007).

3.6 Summary

The fundamental and essential theories which support the development of the designed system were discussed in this chapter. It covered major aspects but not absolutely every one. The nascent and formation mechanism of the ECG signal, as well as the analysis and representation of its waveform are portrayed in detail. In the course of the signal conditioning and processing, the section described the amplification criteria, the principle of Fourier analysis, techniques in both analogue and digital filtering as well as basics in ADC. The communication protocols which support the signal transmission in the networks, e.g. the Ethernet, the Internet and WLAN, were introduced, starting from the OSI model, TCP/IP model, HTTP and closing with IEEE 802.11 specification.

CHAPTER 4

System Design

4.1 Overview

Much of the landmark research in the telemedicine field was reviewed in the last chapter. These works are so important that the exploration and development of the research evoke inventive proposals for the telemedicine systems. On the other hand, it must be understood that not all proposed scenarios or identified solutions have practical implementations, or arouse sufficient interest to enable the development of commercial products. However, the current research efforts spread the perspective of the field and make us contemplate beyond the existing scenarios and solutions.

The purpose of this research programme was to develop a novel technique, involving both hardware and software, for an Internet-based digital signal processing system in the application of telemedicine. This concept embeds Ethernet control into the real-time processing system, which is expected to bring innovation to the telemedicine world. This scheme obviates the need for bulky equipment/computer/laptop/smart phone/PDA as the gateway, and avoids unnecessary communication, making the design innovative compared to existing approaches.

To demonstrate the concept of globally accessible monitoring, a distributed application structure should be employed since multiple users, namely doctors, need to require the data from any place in the world via the Internet. Thus, a client-server model is adopted as the main architecture of the overall system. On the patient side, the hardware and software platform which has the functions of acquiring, processing and transmitting the signals; this system is therefore considered as the server. The applied computer (doctor side) with either a web browser or LabVIEW programme is treated as the client. The advantage of the client-server model enables the system to distribute the real-time biomedical information of the patient among several users through the Internet. The illustration shows the core architecture of the system in Figure 4-1.

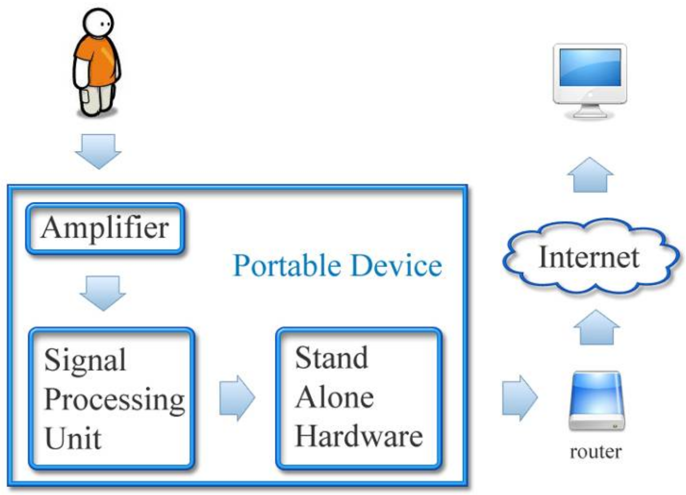


Figure 4-1 Main architecture of the overall system

The inherent characteristics of telemedicine systems require comprehensive solutions, integrating a range of technologies according to different application circumstances. The design methodology and challenges of this telemedicine system are intended for practical purposes, which include the management of multiple biomedical signals; suppression and elimination of noise; comfortable wearable technologies; mobility for the users; universal accessibility to patient's data; conservation of power consumption; personalised and secured monitoring; reliability of monitoring.

To address these requirements, this research concentrated on the development of a high-performance system with connectivity to the Internet, enabling the acquisition, transmission, storage and display of the real-time biomedical signals, mainly ECG signals. Heart rate, respiration and body temperature signals were also developed to prove and assess the system's multi-channel capability. A stand-alone board was built to enable full communication across the Internet. This can also obtain real-time data and be remotely monitored from either a web site or a LabVIEW programme over the Internet. The full functionality of the system is visible to the Internet-based clinicians who are able to obtain signals of patient's body in real-time from the truly portable hardware.

This system platform incorporated six integrated modules. They are: a multiple channel module for detecting diverse biomedical signals; a signal conditioning module for processing raw data; a data transmission module for connectivity to the Internet; a wireless adapter module for ambulatory application; a power supply module for real portability; a storage and data plotting display module for signal reappearance. The block diagram is shown in Figure 4-2.

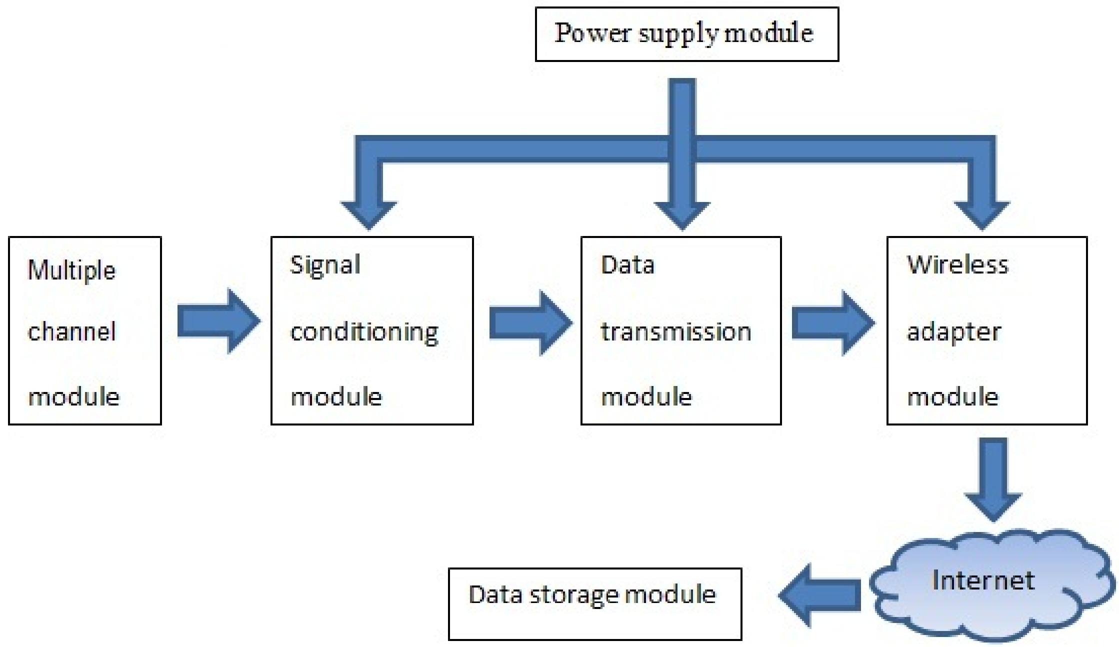


Figure 4-2 Functional block of the designed system

It is worth emphasising that successfully obtaining the biomedical signals in real-time over the Internet is the primary purpose of the design. Using woven/dry ECG sensors must require designing complicated DSP algorithms for diagnostic biomedical signals, which is not principal to achieve the goal. For that reason, the prototype of the system does not make use of woven, dry electrodes which are still an emerging technology and prone to severe artefact. Nonetheless, the flexibility and modularity of the framework facilitates rapid reconfiguration, allowing it to be used with the new generation textile sensors incorporated into intelligent clothing.

4.2 Multiple-sensor module

The multiple-sensor module is the interface connecting the rest of the system to the patient's body. This module is designed to support diverse signals detection in multiple biomedical disciplines, multi-channel monitoring, and offering wearable and comfortable performance to the wearers. Modern telemedicine systems entail an assortment of biomedical signals monitoring as heterogeneous information with medical interest is required by clinicians for diagnoses. At this point, multiple channels have been developed to monitor ECG, heart rate, body temperature and respiration signals in real time. More channels may be expanded by similar means, depending on the capability of the transmission channel and the ADC.

4.2.1 ECG signal and heart rate detection

The designed system primarily monitors Lead I and/or Lead II ECG. ECG detection details concerning Lead I and II were considered in the last chapter. Two Ag/AgCl electrodes with a pair of shielded ECG leads are employed to detect the raw ECG signals and pass them to the signal conditioning module. The ECG leads are arranged as twisted wire pairs. In general, the more tightly wound the twisted pair cable, the more effective is the cancellation of common-mode interference.

Heart rate is an important clinical parameter in the determination of a subject's cardiac health and is expressed as the number of beats per minute. Accurate estimation requires the signal to be relatively noise free since typically the rate is evaluated or calculated by the measurement of consecutive R peaks in the ECG signals. The interval between two adjacent R peaks, so-called R-R interval, provides the instantaneous heart rate, the inverse of which yields heartbeats per second (Electrophysiology., 1996). Figure 4-3 shows a sketch of the R-R interval in the ECG signal.

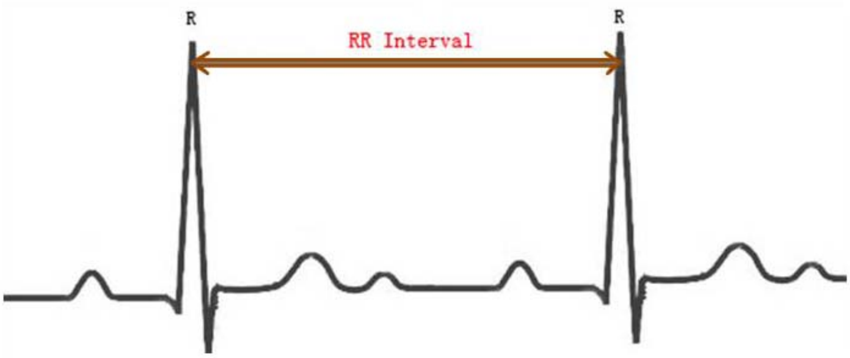


Figure 4-3 The R-R interval in the ECG signals

The algorithm to measure the R-R interval detects two successive R peaks and calculating the value of temporal interval. In clinical monitoring, the heart rate is more commonly given in beats per minute (BPM) and is calculated using:

$$HR = \frac{1}{RR_{Interval}} \times 60 \tag{4-1}$$

Where HR is the heart rate, $RR_{Interval}$ is the temporal R-R interval. The algorithm of the measurement is depicted in Figure 4-4.

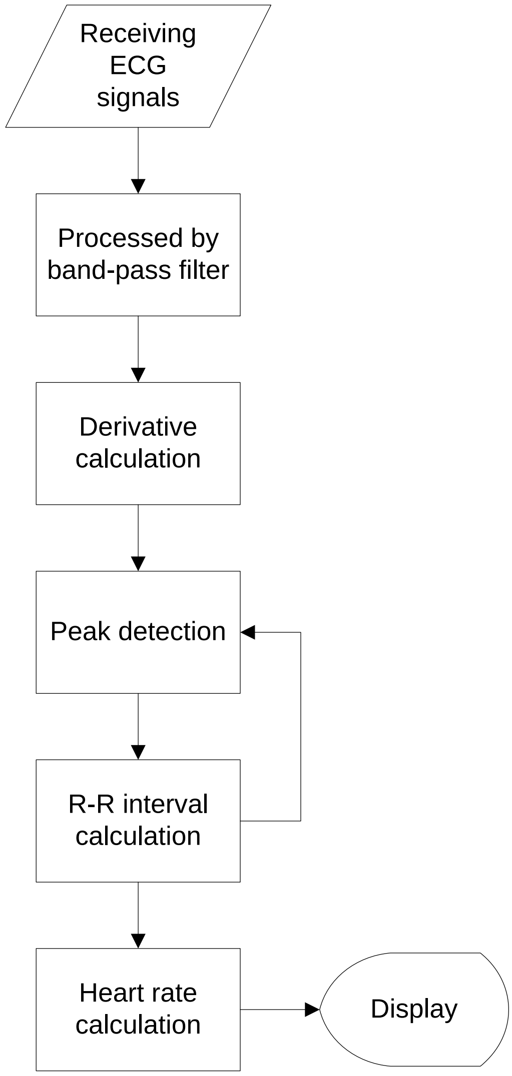


Figure 4-4 The algorithm of heart rate measurement

The heart rate detection algorithm in this work was implemented using LabVIEW (Version 2010) from National Instruments^{TM}. Figure 4-5 shows the functional layout of the heart rate detection programme in LabVIEW.

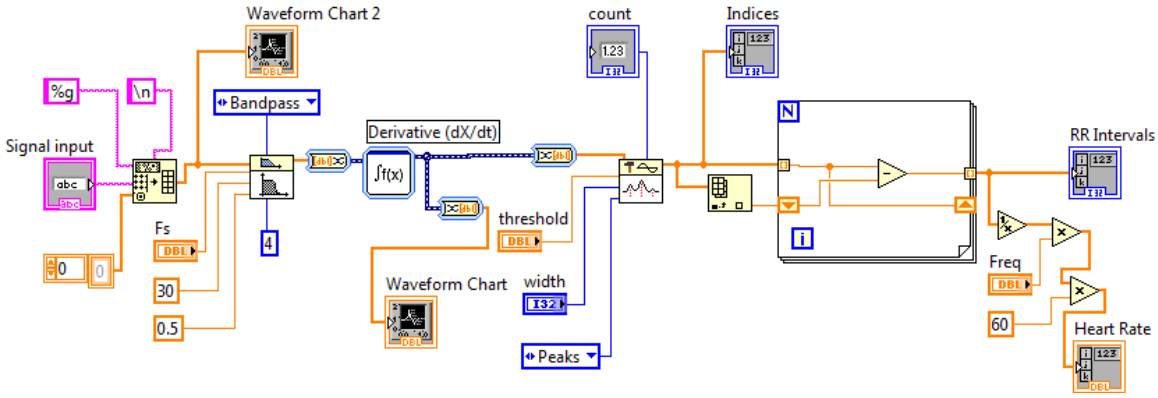


Figure 4-5 LabVIEW programme layout of heart rate detection

The peak detection function in the programme tests each peak from the ECG data against a set threshold magnitude. Because the R peaks are the highest ones in the ECG signals, any peak lower than the R peak signal are ignored by the functions. The locations or indices of the detected peaks are obtained in order to calculate the R-R intervals and the heart rate using Equation 4-1.

4.2.2 Respiration signal

Respiration is another main vital sign in clinical monitoring. In physiology, it refers to the transport of oxygen from the atmosphere to the cells and the transport of carbon dioxide in the opposite way. One of the commonly used methods to monitor respiratory rhythm is to measure changes in thoracic or abdominal circumference, which is in turn caused by changes of gas volume in the lungs (Webster, 1998).

A piezoelectric based respiration belt transducer is employed in this work to measure the respiration signals. A commercial product Pneumotrace II model 1132 made by UFI is adopted. It includes a piezoelectric based sensor strap and an adjustable belt. A snapshot of the transducer is shown in Figure 4-6.

Figure 4-6 Pneumotrace II model 1132 respiration belt transducer

The strap of the transducer is elasticated, whereby the fluctuation in girth of the thorax or abdomen induced by breathing can be detected. The belt is attached to the sensor strap and may be adjusted to fit patients of different sizes. A positive voltage, in the range between 20 mV and 400 mV, is generated by the transducer. Since this transducer generates a small voltage and is active, i.e. it does not consume external power, it meets the safety requirement for low power portable applications. In addition, this piezoelectric transducer has both good linearity and sensitivity (4.5 ± 1 mV/mm) (ADInstruments, 2011). The raw signals are de-noised by an analogue low pass filter with a cut-off frequency of 100 Hz. After appropriate amplification with a gain of 2, the output of the transducer passes to an ADC configured within the range of the sensor system's output. The circuit layout of respiration sensor conditioning system is shown in Figure 4-7.

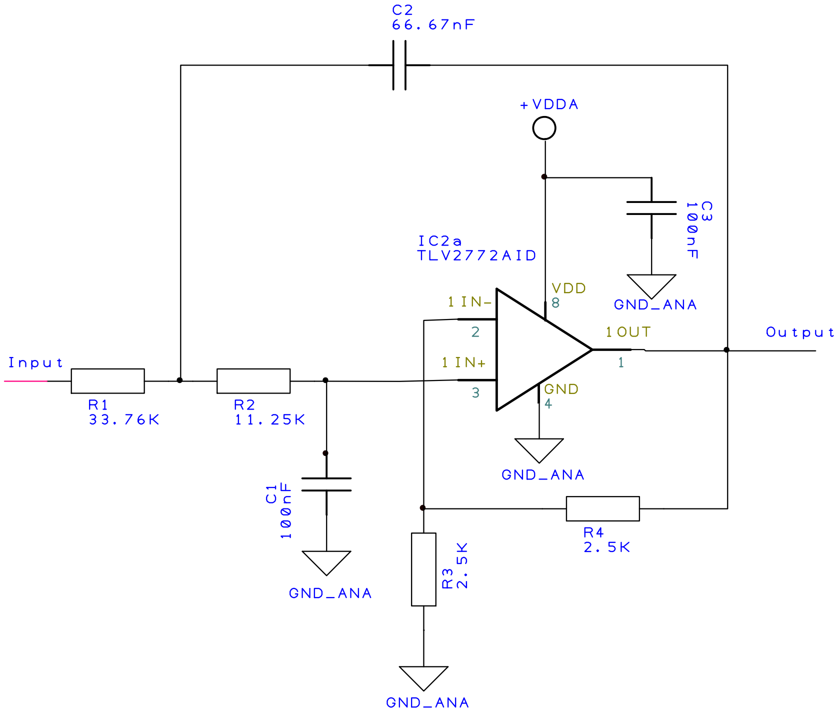
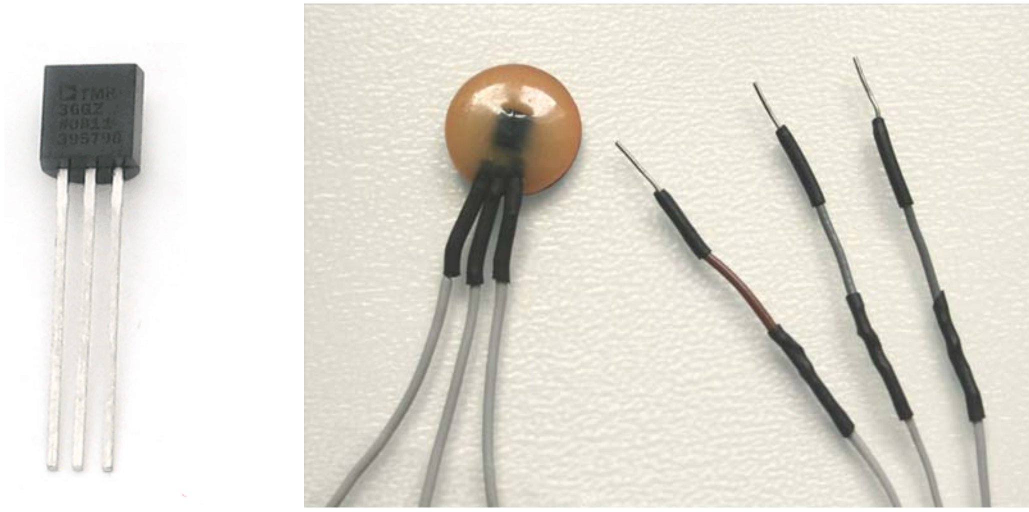


Figure 4-7   Respiration signal conditioning circuit

4.2.3 Body temperature

In a clinical setting, a patient's body temperature is another vital sign which often requires monitoring. In the context of this research, body temperature refers specifically to skin temperature. Normally body temperature fluctuates over the course of the day, with the change of seasons and under different circumstances such as activity levels of the patient. It is also varies across the body, both externally and internally. Human body temperature ranges from 30 °C to 40 °C and the normal temperature is 37 °C, varying by about 0.5 °C during the day (Boano et al., 2011). Various methods can be used to measure body temperature, such as body cavity measurement, oral measurement and skin-based measurement. In this study, skin temperature measurement was undertaken for the purpose of assessing the multiple sensors/channels functionality of the designed system. The human body uses the skin to regulate core body temperature, which in turn impacts on the temperature of the skin via sweating, vasodilatation or vasoconstriction.

To detect the skin temperature, an Analog Devices™ TMP36 precision centigrade temperature sensor was used. This generates a voltage output which is linearly proportional to the temperature. The TMP36 has a wide measurement range from -40 °C to 150 °C with relative low self-heating. It was also selected also because it does not require a negative voltage supply, simplifying the design. The TMP36 has an output range from 0.1 V to 2 V, with a DC offset of 500 mV. This sensor is intended for single supply operation from 2.7 V to 5.5 V, which here was provided by a 3.3 V supply, taken from the front-end analogue module. Figure 4-8 (a) shows a snapshot of the 3-lead TMP36 (TO-92 package) (Analog Devices, 2010b).



(a) *(b)*

Figure 4-8 (a) Snapshot of the 3-lead TMP36 temperature sensor, (b) The skin-based temperature probe

Based on 3-lead TMP36, the skin-based temperature probe was made in the laboratory to interface to the calibration circuit and the ADC. Figure 4-8 (b) shows a photograph of the skin-based temperature probe. The probe is composed of a good thermal-conduction copper disc and a 3-lead TMP36. The TMP36 sensor head is attached and sealed on one side of the disc by thermally conductive, waterproof epoxy potting compound with good electrical insulation characteristics. The copper disc is 17 mm in diameter and less than 0.5 mm in thickness to ensure rapid heat transferral. Its smooth surface ensured good skin contact. Three leads extended from the TMP36 connect to the 3.3 V power supply, ground and signal output. The temperature, in degree Celsius, can be obtained from the output voltage, using the following formula (Analog Devices, 2010b).

$$T = \frac{V_{output} - 500}{10} \tag{4-2}$$

Where T denotes measured temperature in Celsius and V_{output} is the output reading of the TMP36 in millivolts.

According to the typical human body temperature range (30 ˚C to 40 ˚C), the output of the temperature probe will not exceed upper limit 900 mV and lower limit 800 mV. In order to maximise the dynamic range available from the ADC, the voltage output from the sensor was amplified by a gain of two. The temperature value read from the 12-bit ADC is given as:

$$T_{digital} = \left(\frac{1}{2} \times V_{ADC} \times \frac{3300}{4096} - 500 \right) \div 10 \tag{4-3}$$

Where $T_{digital}$ is the temperature value obtained via the software in Celsius and V_{ADC} denotes the ADC's reading.

4.3 Signal conditioning module

All biomedical signals including the unprocessed ECG signals must first be processed in several ways before being transmitted, displayed and stored. First, the voltage amplitude of the raw ECG signal is in the level of millivolts, so gain must be applied. Second, the signals are often contaminated by an assortment of interference which may stem from movement artefact, electrode contact impedance change, respiration, power-line interference, EMI from other electronic devices and so on. Third, the signals should be digitised for further processing, transmission and storage. However, the amplitude of the raw signals is usually out of the ADC input range. To address the three issues discussed above, there are corresponding sub-modules to process the analogue signals, shown in Figure 4-9.

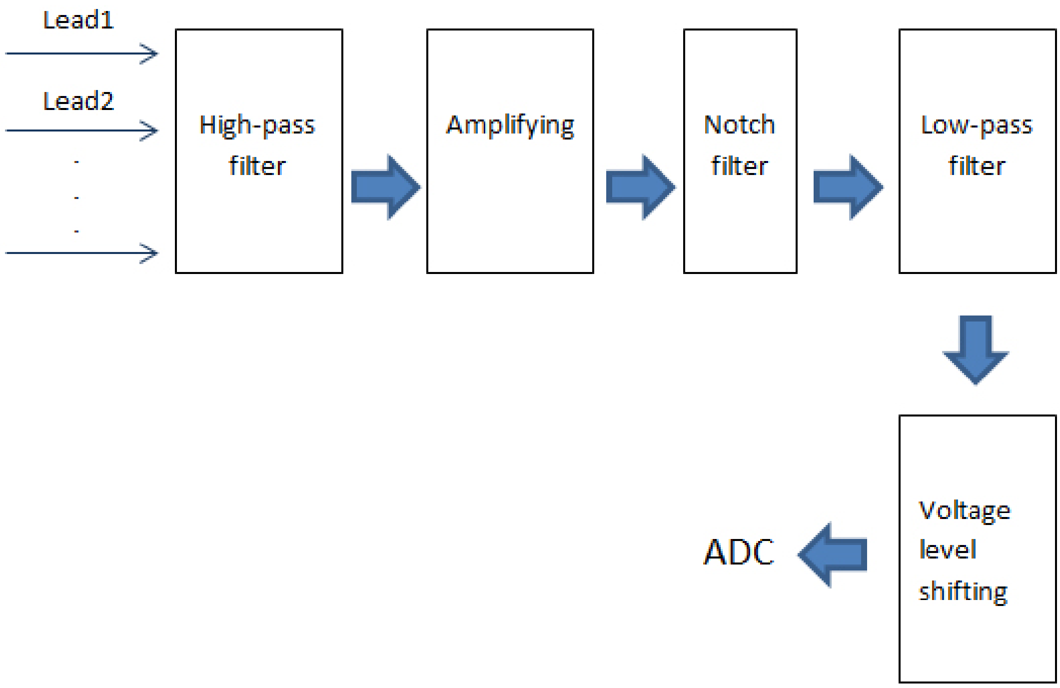


Figure 4-9 Functional block of front-end signal conditioning module

Amplification is an essential step necessary for reproduction of the signals. After that, a combination of filters including high-pass, low-pass and notch filter are required to remove noise over different frequency ranges. Level shifting circuitry is also often necessary to adjust the voltage offset of the signals to lie within the ADC input range. This front-end module represents the preliminary processing stage for the analogue biomedical signals.

4.3.1 Amplification

Amplification of ECG signals is necessary for the reasons given above. However, the amplitude of the common-mode noise normally exceeds that of the ECG signal, so common mode noise must be minimised appropriately.

To resolve this noise issue, an accurate analogue instrumentation amplifier is used to amplify the ECG signals and to suppress the common-mode noise concurrently. As the ECG signal is differential, the most appropriate amplifier is a differential-based type. The major considerations of amplifier selection are the requirements of low power consumption, low input-referred noise, low input bias current, high differential gain, low common-mode gain, high input impedance as well as CMRR.

As discussed in section 3.3.1, the input impedance of the amplifier ought to be much larger than the source impedance in order to reduce loading of the signals. The input impedance is typically recommended to be not less than 10 MΩ (Webster, 1998). Since the amplitudes of the original ECG signals only have a couple of millivolts, an amplification of 1000 times is reasonable to regulate the voltage amplitudes to the level of volts. As common-mode noise is a major source of interference for such signals, an instrumentation amplifier with high CMRR can minimise the effect of common-mode voltages. The value normally should be higher than 90 dB from DC to 60 Hz. As the half-cell potential differential appears as a DC input offset voltage, the variable DC offset of up to ±300 mV may add to the inputs of the amplifier. The conditioning circuitry should tolerate such a high value to avoid saturation. The table 4-1 is a summary of major performance requirements for a typical ECG signal amplifier.

Table 4-1 Major performance requirements for a typical ECG signal amplifier

Parameter	Value
Input range of signal	±5 mV
Input impedance	10 MΩ or greater
Gain range	1000 or greater
Minimum CMRR at ECG frequency	90 dB (DC to 60 Hz)
Minimum tolerable input DC offset	±300 mV

Based on these major requirements, the performance of the amplification module can be achieved by using a high accuracy instrumentation amplifier such as the AD620 by Analog

Devices™. This is supplied as a small 8-lead monolithic SOIC or DIP, and has a lower power consumption, typically requiring a supply current of 1.3 mA. This instrumentation amplifier can offer gains of 1 to 10000 using a single resistor. 100 dB minimum CMRR can be achieved with corresponding gains. Moreover, the low input noise and low input bias current performance of the AD620 avoid bringing incidental inference into the original signals. These specifications make the AD620 appropriate for wearable/portable/remote precision ECG signal acquisition systems. The circuit block depicts the design scheme of the amplification module in Figure 4-10.

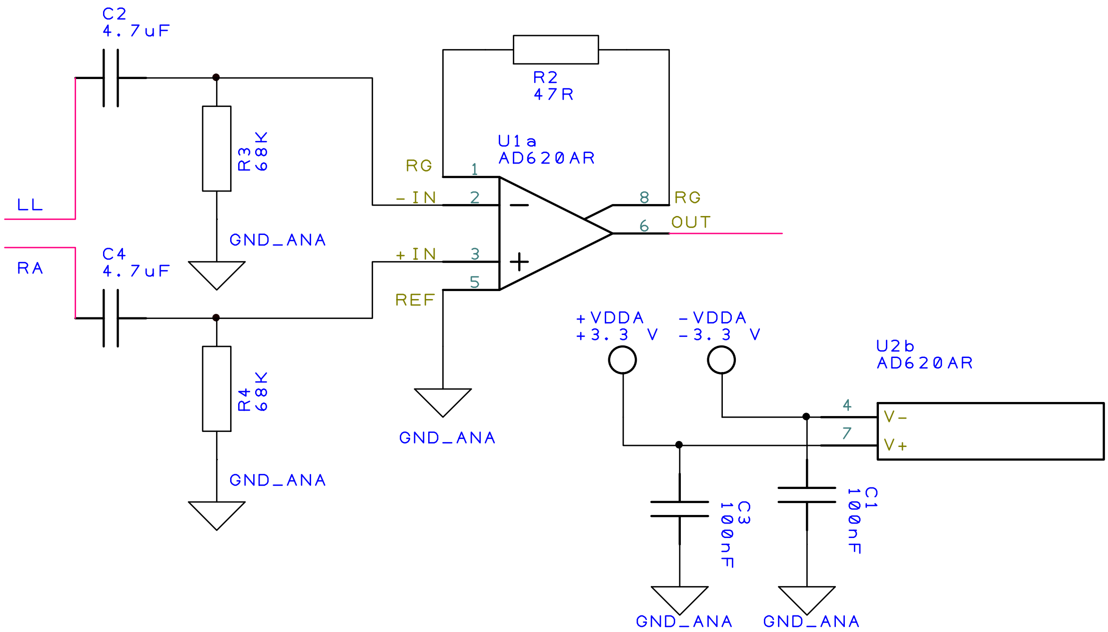


Figure 4-10 Design scheme of the amplification module

In this work, a single 47.5 Ω resistor is required to set a gain of 1000, which correspondingly attains a CMRR of 110 dB to 130 dB over a frequency range of 150 Hz, according to typical performance characteristics shown in Figure 4-11 (Analog Devices, 2004).

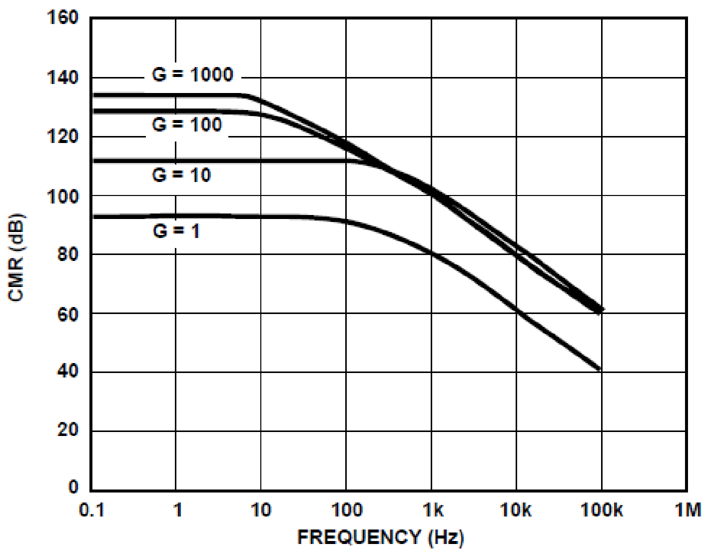


Figure 4-11   Typical CMRR over frequency

The power supply operates at Vs = ±5 V DC. Accordingly, the typical input voltage range is from –Vs+1.9 V to Vs–1.2 V, to be precise, –3.1 V to 3.8 V here (Analog Devices, 2004). Therefore, the raw analogue ECG signal inputs (in mV) will surely fall into this scope. Moreover, environmental noise can affect the power supply voltage for the amplifier as a result dispatching the effect into the circuit. It can be minimised by means of shunting the environmental noise with 100 nF decoupling capacitor which is deployed as close as possible to the amplifier at the power supply terminals.

Although the amplifier has good CMRR, other forms of noise are still present in the signal after amplification, and these must be removed by filtering, as discussed below.

4.3.2 Multiple filtering

After amplification, multiple analogue filters are applied to recover the ECG signal. Various kinds of noise contaminate the ECG signals, which derive from various sources. There may be variable contact between the electrode and the skin. Movement artefacts also cause changes of the electrode-skin impedance. Normal human biophysical motions like respiration can evoke baseline drifting. Power-line interference introduces 50 Hz (UK) or 60 Hz (USA) noise. The ECG Lead cables serve as antennae and detect high frequency EMI generated by other electronic devices. These interferences cause noises in the different frequency harmonics. The following table elaborates on the noise sources and the corresponding frequency domains.

Table 4-2 Noise sources and their corresponding frequency domains

Noise source	Corresponding frequency domain
Variable electrode contact	Baseline drifting, close to zero Hz
Movement artefacts	Baseline drifting, close to zero Hz
Respiration	Baseline drifting, close to zero Hz
Power-line interference	50Hz (in the UK)
EMI	Broadband and RF high frequency

According to the table above, the frequency spectrum of the noise spreads over the signal bandwidth. With the purpose of meaningful extraction, the frequency range of the ECG signal should be examined precisely. Generally the ECG signals have frequency range approximately from 0.05 or 0.5 Hz to 100 or 150 Hz, which contain the required cardiology information or the fidelity of the recorded ECG signals. For different purposes, several criteria dictate different interpretation. With the intent of visible monitoring, it requires frequency range within 0.5 to 35 Hz. In diagnostic analysis mode, it has wider frequency range requirement – from 0.05 to 100 Hz.

The ECG signal is characterised by six principal segments, named P, Q, R, S, T, and U as shown in Figure 4-12.

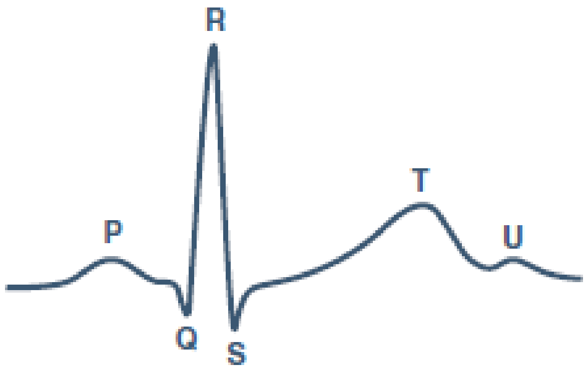


Figure 4-12 Portrait of ECG signal segment P, Q, R, S, T and U

The precise values of the roll-on and roll-off of the pre-conditioning filters are important since they can affect considerably the shape of the filtered signal and thereby any interpretation. The lowest examined cut-off frequency should not exceed 0.5 Hz. The T-wave contains a harmonic component around 1.0 Hz in frequency spectrum. The peaks of the Q, R, and S segments which are of valued clinical information contain frequency spectrum above 30 Hz up to 100 Hz. ECG signal components with frequency harmonics beyond 100 Hz, like the sharp peak of R-wave, have less important for observational interpretation. Accordingly, analogue high pass and low pass filters are employed to limit the frequency scope of the analogue biomedical signals. This diagram in Figure 4-13 illustrates the topology of the signal components and the noise in the frequency spectrum.

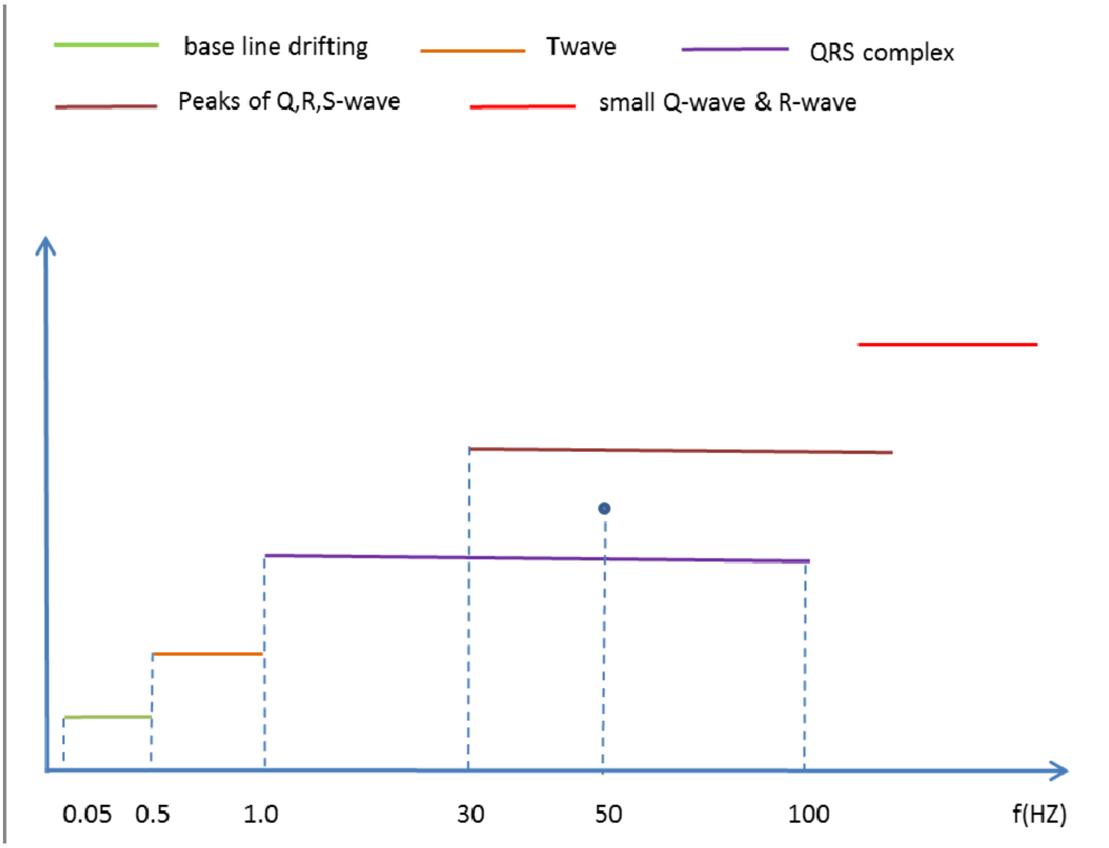


Figure 4-13 Diagram of the frequency spectral topology of signal components and noise

If the analogue high pass filter is designed at a cut-off frequency of 0.5 Hz, it may lead to distortion of the ECG signal's low frequency components like the ST segment. To reduce the distortion of the ST segment, the American Heart Association recommends that the low cut-off frequency should be 0.05 Hz for diagnostic ECG. However, while a wider frequency range preserves the fidelity of the signals, baseline drifting caused by multiple sources can affect the coherent alignment of the signal's baseline. A little deviation of the ST segment of the ECG signals can be tolerated in the application when the lower cut-off frequency was 0.5 Hz in this design.

In summary, with the aim of monitoring the biomedical signals accurately, the cut-off frequency of analogue high pass filter was designed at 0.5 Hz, and the cut-off value of analogue low pass filter was set at 100 Hz. A notch filter at 50 Hz was also included to minimise mains interference.

With high gain application even a small dc voltage offset at the input of the amplifier can result in output saturation. As discussed, the normal-mode DC offset may reach as high as ±300 mV but the maximum input of the ECG signal is only ±5 mV. If using one-stage amplification without preconditioning, the output value of the signal with the gain 1000 may cause output saturation. To avoid that, prior to the signal amplification, it is primary to eliminate the large DC offset from the raw biomedical signals. Capacitive coupling is applied to let the AC signal pass and block the DC component. Along with a ground resistor, the analogue high pass filter with a cut-off frequency 0.5 Hz is utilised to accomplish the task. Figure 4-14 is the design scheme of the analogue high pass filter circuit.

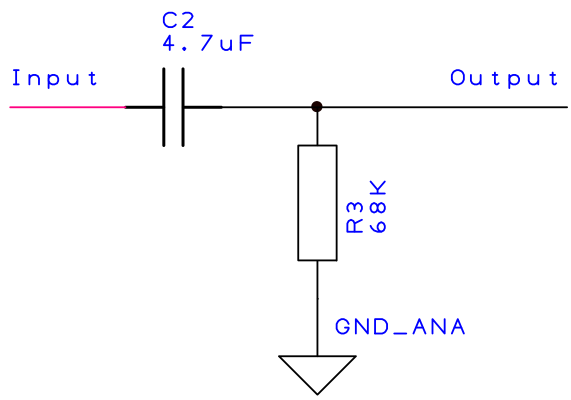


Figure 4-14 Design scheme of analogue high pass filter with cut-off frequency 0.5 Hz

Dual 1st order RC high-pass filters were used on each input of the lead. The low frequency components below 0.5 Hz are abolished or attenuated by the analogue high-pass filter. Thus, any large DC offset is eliminated and base-line drifting is restrained. After pre-filtering, the ECG signals are fed into the instrumentation amplifier.

Actually, the CMRR of the instrumentation amplifier is in fact a finite value, which means it merely cannot suppress entirely common-mode noise. The noise induced by power-line interference at 50 Hz can still contaminates the ECG signals, particularly the P-Wave and T-Wave parts (Chaudhuri et al., 2009). To reduce the power-line interference, a band-stop or notch filter at central frequency 50 Hz is employed next to the output of the instrumentation amplifier. The circuit design of the filter is given in Figure 4-15.

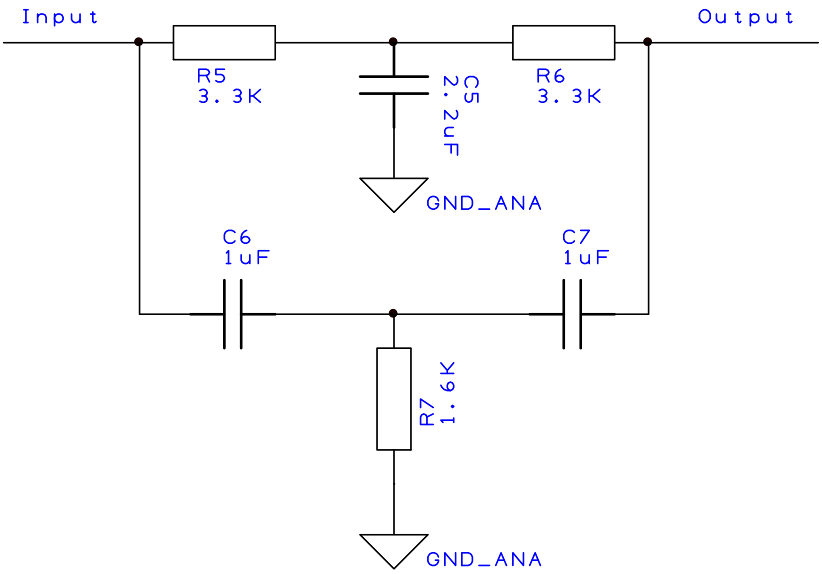


Figure 4-15   Design scheme of analogue notch filter at 50 Hz

This analogue RC passive circuitry is a twin-T notch filter. Good matching of the components is required to achieve a good null at central frequency 50 Hz. Although the frequency response at 50 Hz attenuates quickly, the response at two sides of the cut-off frequency may fall away slowly and affect some bandwidth. However, for monitoring, the performance of the notch filter is adequate to maintain an acceptable fidelity of the ECG signal.

ECG signals can also be influenced by environmental noise from electromagnetic radiation. For instance, the lead wires and patients serve as antennae and pick up the radio frequency interference (RFI or EMI), which degrades the signal of interest. It is fortunate that upper frequency limit of the ECG signal is approximately 100 Hz. To counter the EMI effect, a low pass filter with a corner frequency 100 Hz can be employed without loss of biomedical information. The circuit design of an analogue low pass filter is shown in Figure 4-16.

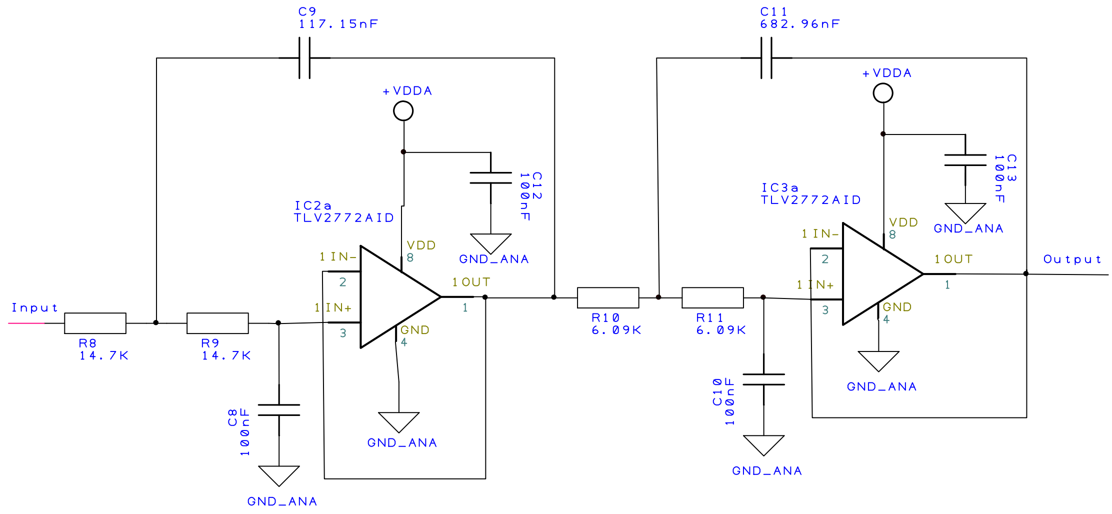


Figure 4-16   Design scheme of analogue low pass filter with cut-off frequency 100 Hz

This analogue active low pass filter has Sallen-Key architecture. The implementation of the 4th-order Sallen-Key filter uses an operational amplifier with 0 dB gain in two stages. Hence, the circuit is appropriate for this gain sensitive application.

4.3.3 Voltage level shifting

The use of multiple filters provides such significance on the reappearance of the ECG signals. In order to facilitate further processing, transmission and storage of the signals, the analogue version has to be converted to digital format. The input voltage range of the used ADCs typically swings from ground to +3.3V. The output of the front-end signal filtering circuit should accommodate the scale of the adopted ADC's specification. To do this, a voltage level shifting circuitry helps to calibrate the signals to a proper level for the ADC without changing the peak-to peak voltage of the signals. The circuit layout of the level shifting scheme is shown in Figure 4-17.

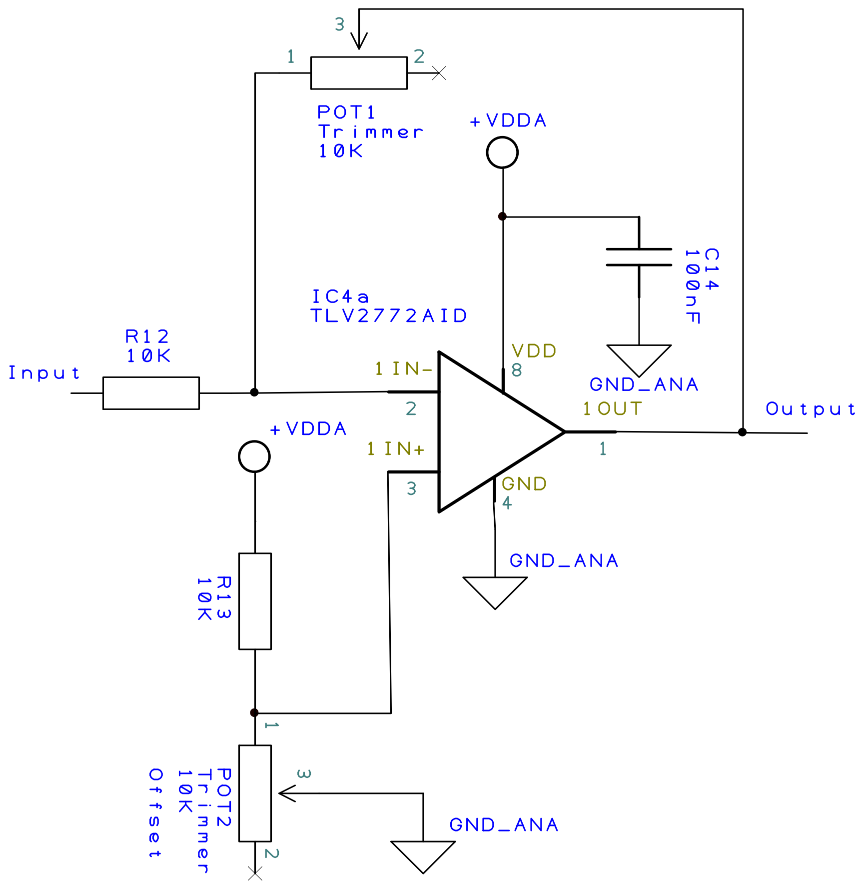


Figure 4-17 Design scheme of voltage level shifting circuit

This voltage level shifting circuit use a differential operational amplifier to alter the voltage level. Potentiometer POT1 adjusts the peak-to-peak amplitude of the signals and potentiometer POT2 shifts the baseline to fit the typical input range of the employed ADC. Therefore, as long as the adjusted gain and voltage offset have been fixed on the proper values by proper adjustment of these two potentiometers, the input signals dispatched to the ADC can be converted appropriately.

4.3.4 PCB implementation of signal conditioning module

The signal conditioning unit was designed as a 110 mm × 77 mm PCB board. The ECG leads connect to the male panel sockets and the outputs are fed to the data transmission module. The following picture in Figure 4-18 shows the PCB board of the signal conditioning module.

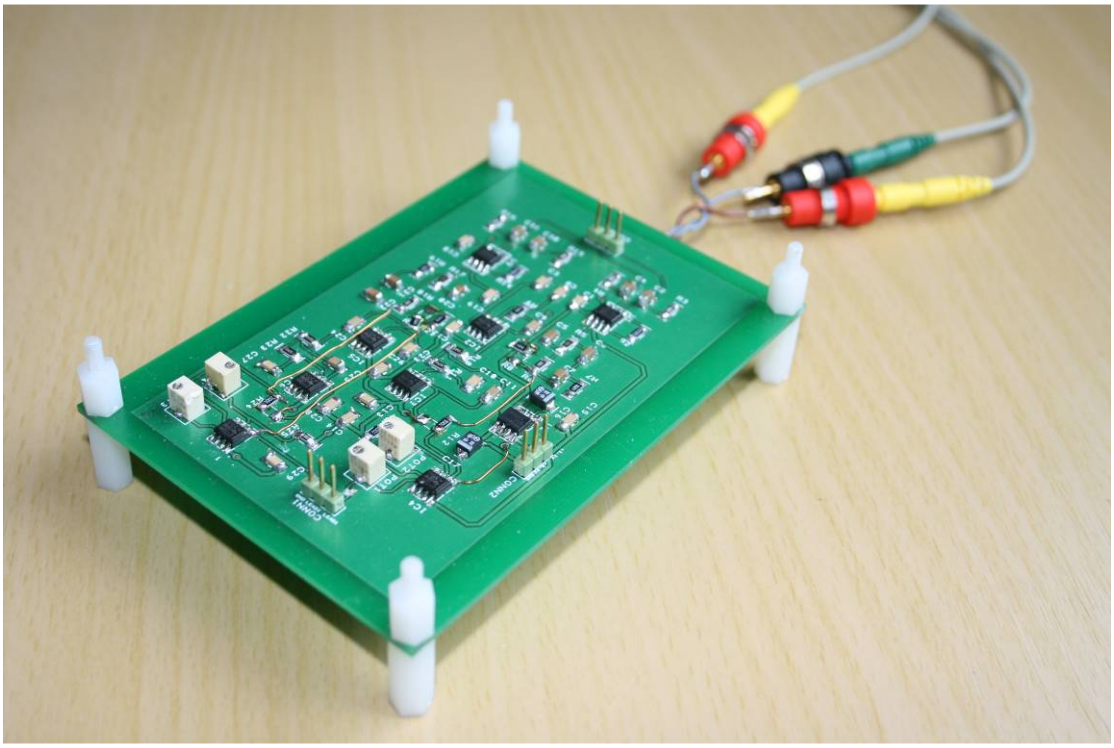


Figure 4-18 PCB board of the signal conditioning module

4.4 Data transmission module

The core of the whole system is the data transmission module which relays the biomedical signals to the Internet in real-time. This module provides a novel solution to replace the conventional control masters or transmission gateways such as the computer, laptop or smart phone, yielding the possibility of extensive applications where the conventional approach would not be appropriate or feasible. It has distinguishing features of being high-integrated, truly embedded and has portable.

To achieve this objective, stand-alone hardware executes the performance that enables the users to obtain the information from the target patient via the Internet. A microcontroller is selected to meet the demand of data acquisition, Internet connectivity, and provides remote monitoring combined with a high resolution display. With a large range of performance and peripherals, the ColdFire portfolio from Freescale™ offers significant flexibility and choice to facilitate the system design. Founded on the microcontroller, there are several functions in the data transmission module to manipulate the signals and establish the Internet connectivity, the functional block of which is depicted in Figure 4-19.

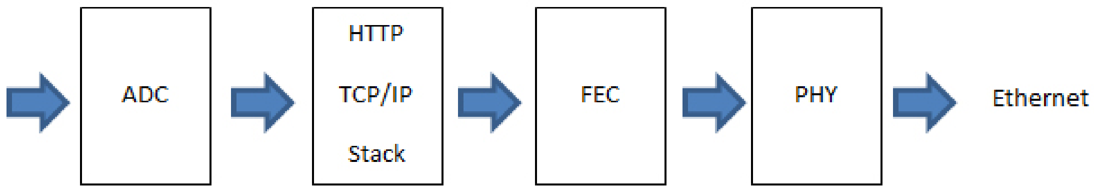


Figure 4-19 Functional block of data transmission module

The analogue biomedical signals are delivered from the signal conditioning module to the data transmission module. First of all, analogue-to-digital conversion is indispensable and the digitised signal is stored temporarily in the RAM of the microcontroller. Then the data are handled and dispatched based on the Internet protocol – TCP/IP stack. The Ethernet connection is provided by the PHY and FEC modules integrated in the ColdFire microcontroller. An Ethernet cable with an RJ45 connector or a wireless connection links the stand-alone hardware with a home router.

4.4.1 Global distribution

Modern computer and digital systems are associated with large amounts of information, which is often transmitted over a network or a mobile telephony system, especially the Internet. The Internet is a worldwide, publicly accessible series of interconnected computer networks, which has the most comprehensive deployment and the best cost performance. The US Census Bureau reported that, as of June 30, 2010, the Internet usage statistics news and population stats show there are over 1.9 billion users in the world. The usage of the

Internet as a communication medium can offer flexible, easy-to-access and cost-effective biomedical monitoring systems which are not restricted to any geographical region.

Associated with the Internet broadband, personal IP address offers individual monitoring rather than multi-user sharing this bandwidth. This provides the advantage of rich biomedical information delivery for a single patient within a limited bandwidth, merely occupying one IP address.

4.4.2 Analogue-to-digital conversion

The analogue signals must be digitised prior to transmission, processing and storage. The sample rate must be chosen appropriately to ensure the fidelity of the ECG signals. Moreover, the cut-off frequency of the low pass filter has an effect on the signal and determines the selection of the sample rate.

According to the Shannon sampling theorem or the Nyquist theorem, sampling must take place at a rate of at least twice the highest frequency component contained in the analogue signal (Gaydecki, 2004). In this case, to be within the correct range, a minimum sample rate of 200 Hz is required, since the cut-off of the low pass filter is 100 Hz.

In practice, on-chip ADCs possess several merits for biomedical device applications (Fuchs et al., 2002, Van Helleputte et al., 2008). They facilitate circuit layout, reducing the number of tracks on the board, resulting in less interference coupling into the system. Reducing the layout size and complexity of the circuitry allows biomedical instrument miniaturisation. In addition, on-chip ADCs reduce the power requirement for portable design.

Based on the design scheme of the system, a multi-channel approach was adopted to enable the recording of heterogeneous biomedical information. Hence, multiple ADC channels should be considered to monitor diverse biomedical signals.

To attain the requirements discussed above, an on-chip eight-channel ADC which is built in the ColdFire microcontroller was configured to digitise the analogue signals. This block diagram shows the ADC module function in Figure 4-20 (Freescale, 2007).

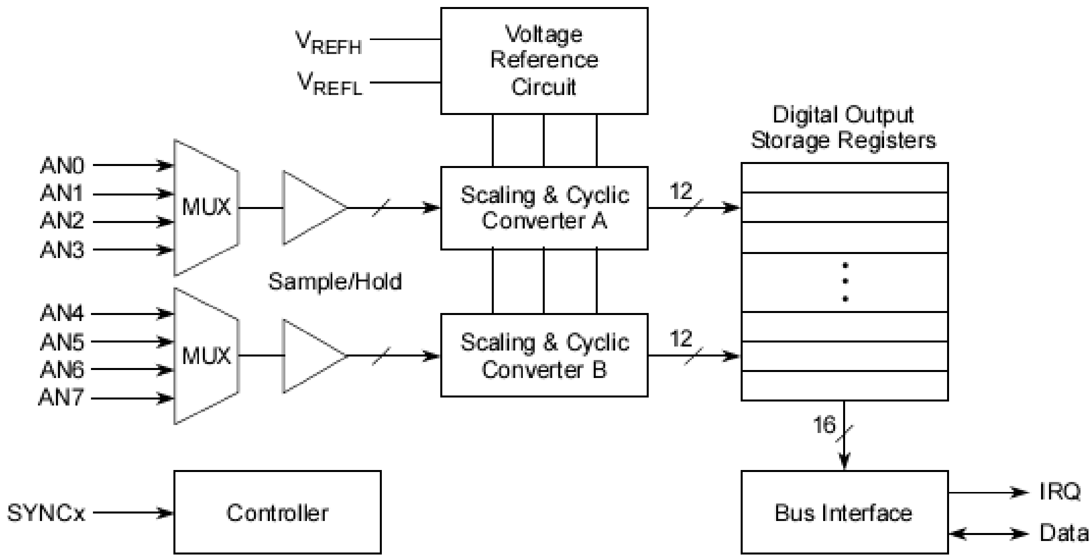


Figure 4-20 On-chip ADC block diagram (Freescale, 2007)

This build-in ADC has 12-bit resolution, which is sufficient for most biomedical signals. It consists of an 8-channel input select multiplexer and two separate and complete ADCs, each of which has its own sample and hold circuits. The two converters share a common voltage reference and common digital control module, which store their results in accessible buffers for further processing. Under the conversion range between ground and +3.3V, the sampling rate for ECG signals is set at 200 Hz. The ADC was configured for loop sequential conversion for long-term continuous monitoring.

4.4.3 ColdFire MCF5223x microcontroller

To fulfil the specific needs of the research, the ColdFire MCF5223x family of microcontroller was selected, allowing Ethernet connectivity in conjunction with real-time DSP. The ColdFire portfolio is targeted at network-based control applications with serial interfaces including Fast Ethernet. They are built on an extensive range of peripheral options which provide multiple performance and connectivity solutions for embedded designs. The excerpted diagram shows the ColdFire MCF5223x Family block in Figure 4-21 (Freescale, 2007).

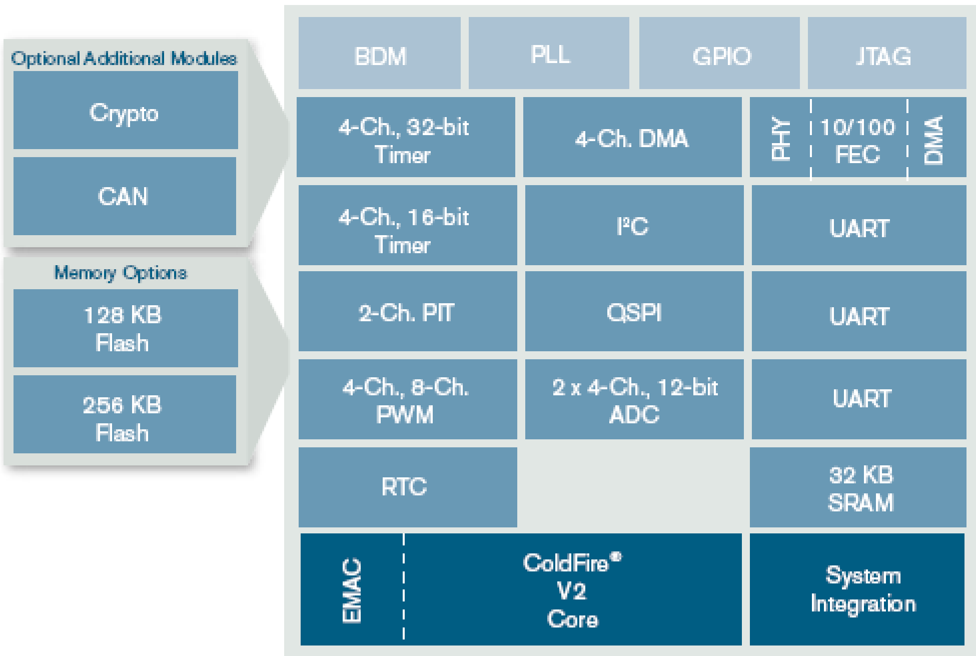


Figure 4-21 ColdFire MCF5223x Family Block Diagram (Freescale, 2007)

Based on the Version 2 ColdFire core, the MCF5223x microcontroller integrates the right set of standard ColdFire peripherals and memory sizes for a compact Internet-enabled platform. On-chip memories comprise up to 256 Kbytes of internal flash and 32 Kbytes of static random access memory (SRAM). Moreover, innovative optional features include a 12-bit ADC, an enhanced programmable I/O state machine, dual-port SRAM capability which enables concurrency, vectored interrupts which enable effective prioritization and real-time control capability, background debug mode (BDM) trace capability which enable write/erase flash and debug, 32-bit timers, a flexible bus configuration and so on (Freescale, 2007). The pre-processed analogue biomedical signals are acquired by the on-chip multi-channel ADC. Subsequently the microcontroller performs signal manipulation and network transmission via the Ethernet port. The circuit schematic design of the microcontroller and its peripherals is depicted in Figure 4-22.

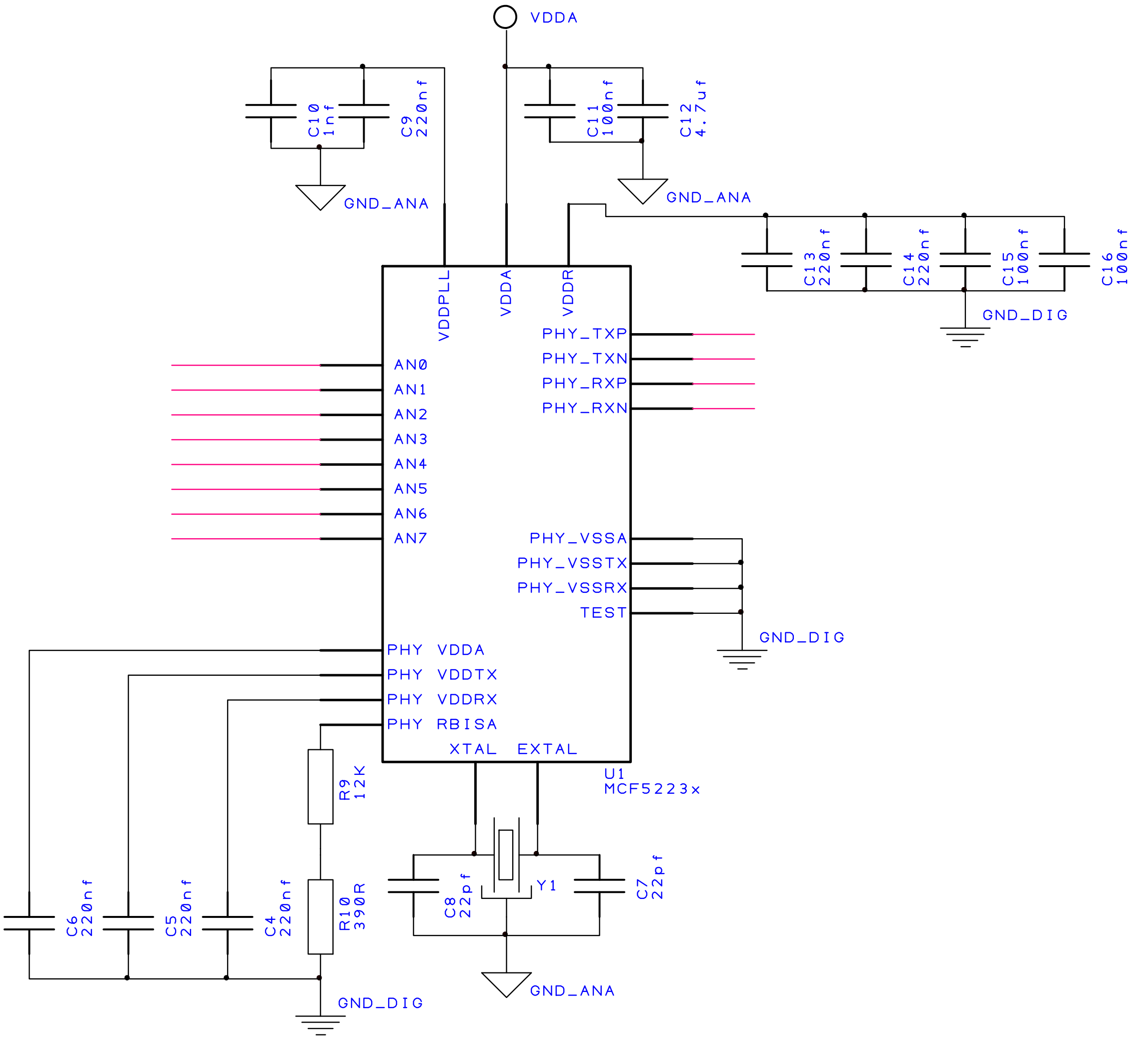


Figure 4-22 Circuit schematic of the microcontroller and its peripherals

The following sections will introduce two key figures of the chip – FEC and PHY modules which perform essential functions of Ethernet connectivity for the system.

Ethernet, as computer networking technology standardised as IEEE 802.3, is the middleware link between the Internet and the networking hardware. An Ethernet port includes two parts: the Media Access Layer which executes a multi-node addressable communications protocol and the Physical Layer (PHY) which represents the mechanical/electrical connection between devices (Walrand, 1998). The MCF5223x microcontrollers are the first with an integrated Fast Ethernet controller (FEC) and Ethernet Physical Transceiver. Those two modules handle the MAC layer and PHY layer respectively.

The Fast Ethernet Controller module is the ColdFire interface to the media (the Ethernet). This module has a high performance Ethernet engine that supports both 100 Mbps (100BASE-TX) and 10 Mbps (10BASE-TX). With a minimum system clock rate of 50 MHz, the FEC supports full-duplex operation (200 Mbps throughput at 100BASE-TX), that is, the packets can be sent and received concurrently by using separate wires for TX and RX (Freescale, 2007).

The FEC is implemented with a combination of hardware and software. The FEC hardware executes all the functions of the IEEE 802.3 Ethernet MAC layer without software intervention, which means that it allows for packet transmission and reception without processor overhead. The advantage of these high level functions in the FEC is decreased software overhead. The software writes the MAC address of the node into the MAC address register to initialize the FEC and the RX/TX buffer rings. For receiving data, the FEC automatically receives, processes, verifies and transfers incoming packets into an RX buffer. For sending data, the FEC is triggered by the software, then automatically transfers the packet from the TX buffer, calculates a Cyclic Redundancy Check, serialises the packet and sends it out to the PHY. The FEC reports a status after the packet is transmitted (Freescale, 2007).

The MAC must be used in conjunction with the on-board transceiver interface and transceiver function to complete the interface to the Ethernet. In the modules of the MCF5223x microcontroller, they are implemented by an Ethernet physical transceiver

(EPHY) and a Medium-independent interface (MII). The Ethernet physical interface module in the MCF5223x chip supports the MII and the MII management interface. It is configurable via internal registers that are accessible through the MII management interface and limited configurability using the EPHY register map. A 25-MHz crystal for its basic operation is required by the EPHY (Freescale, 2007).

The PHY communicates with the MAC layer by the MII. The Ethernet physical transceiver, referred to as the Ethernet physical interface, is an IEEE 802.3 compliant 10BASE-T/100BASE-TX Ethernet PHY transceiver. Integrated Ethernet PHY enables support for data transmission. The block diagram illustrates the EPHY module in Figure 4-23 (Freescale, 2007).

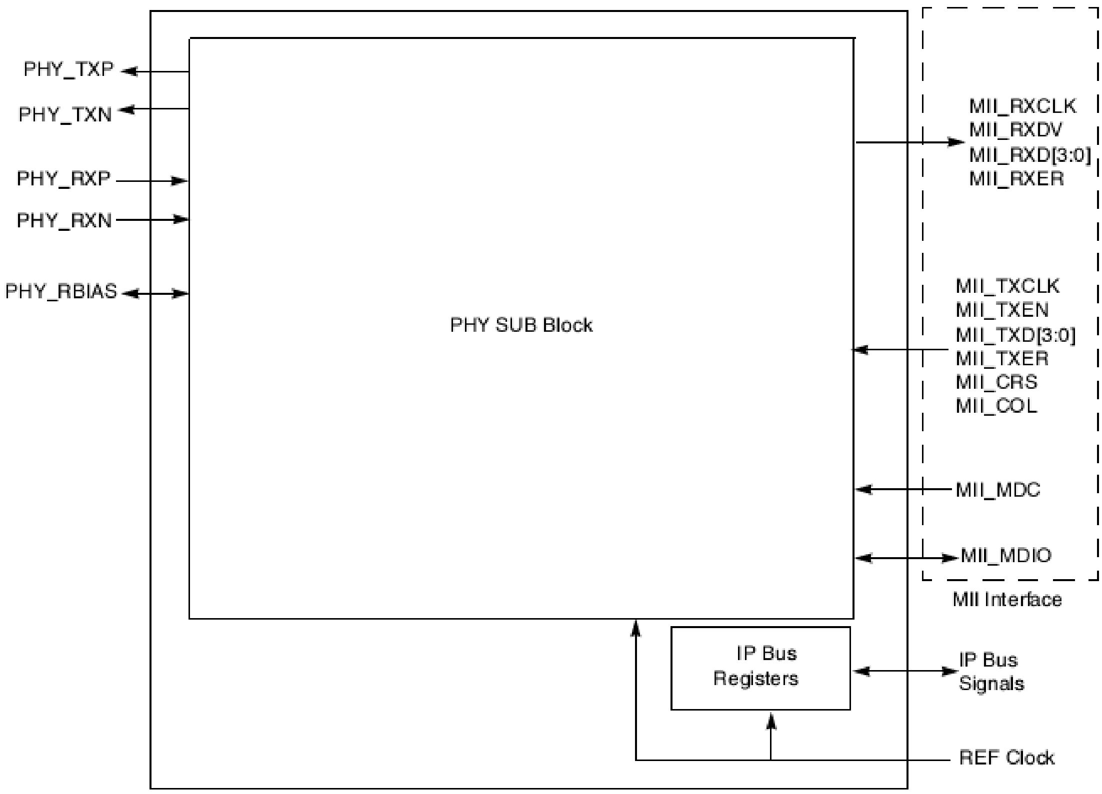


Figure 4-23 EPHY Block Diagram (Freescale, 2007)

100BASE-TX is applied in the research. 100BASE-TX is referred to as the common modern copper physical layer. Commonly, the Ethernet connector in this layer is 8-pin RJ45 connector that is used by most PC networks and broadband. This 8-position 8-contact cable/socket contains eight wires that are grouped in four twisted pairs. Two twisted pairs (PHY_TXP, PHY_TXN, PHY_RXP and PHY_RXN) are used for communications in each direction. The design sketch shows the Ethernet connector circuit in Figure 4-24.

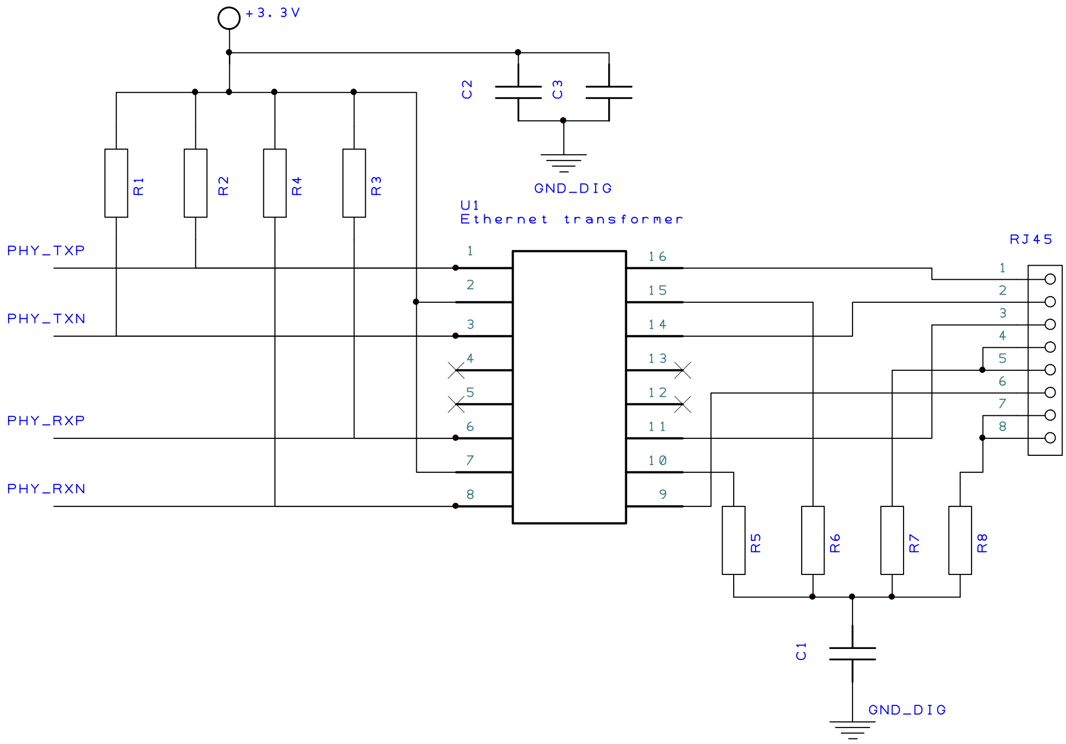


Figure 4-24 Design scheme of the Ethernet connector

A category 5 Ethernet cable with an RJ45 connector links the hardware board to a standard router. The router allows access to the Internet. Thus, at the physical level the designed hardware is endowed with the Internet connectivity over the Ethernet.

4.4.4 PCB implementation of data transmission module

A circuit board is devoted to manipulate the initially processed ECG signals and dispatch them to the Internet. The stand-alone hardware with Internet connectivity undertakes the task of data transmission which provides the possibility of embedded or portable capability. This system communicates with the router using either a cabled or wireless link to transfer the ECG data to the Internet. The photograph in Figure 4-25 shows the data communication PCB board with the signal conditioning PCB board, the ECG leads as well as the Ethernet cable.

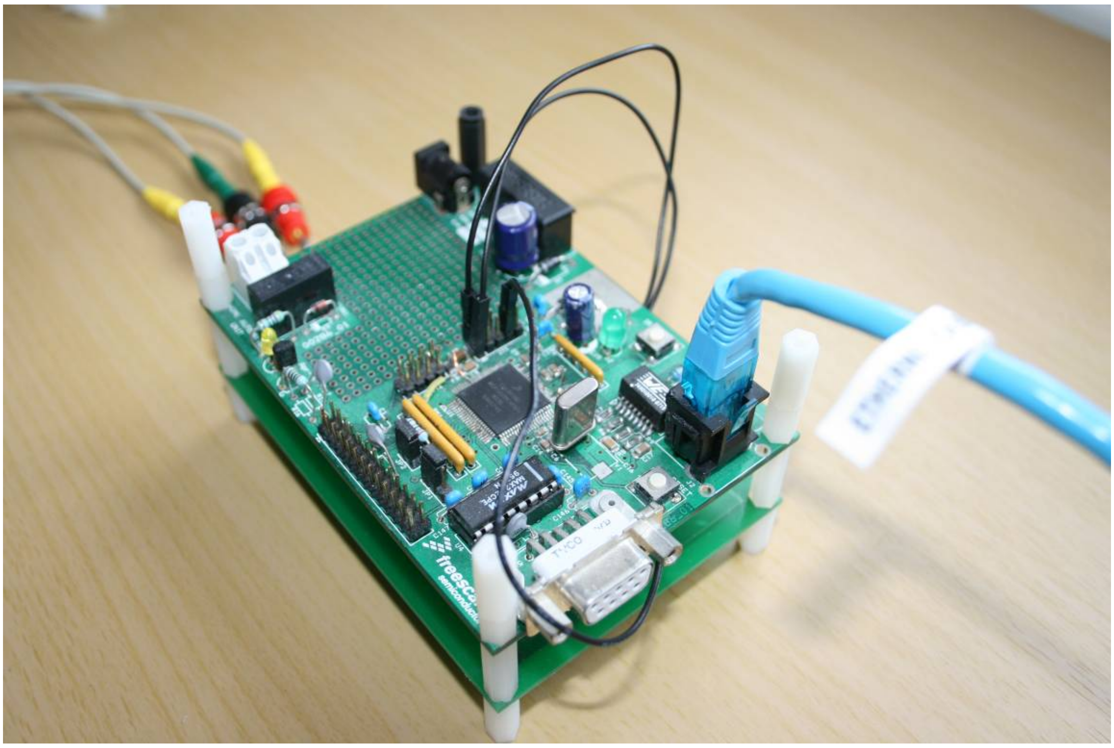


Figure 4-25 Data transmission PCB board with the signal conditioning PCB board, the ECG leads and the Ethernet cable

The data communication PCB board is the same size as the signal conditioning board. It is powered by a mains supply adapter with an output of 5-8 V DC, rated at 500 mA. This board also supplies +3.3 V power to the front-end PCB board. The outputs of the signals from the front-end PCB board are fed to the ADC inputs of the data transmission board. The CAT5 Ethernet cable connects the RJ45 socket and the router. When all connections are in place, pressing the ‘reset’ button on the data transmission board activates the operation of the real-time ECG signal transmission onto the Internet.

4.4.5 TCP/IP stack

The Ethernet can directly provide a simple point-to-point communication mechanism via the data link/MAC layer and the PHY layer, but it cannot establish robust communications such as the Internet. The Ethernet is merely fundamental and necessary middleware for the Internet. The Internet connection is executed by the firmware operating in the microcontroller, the kernel of which is based on the TCP/IP stack.

Defined or referenced by the seven-layer OSI model, TCP/IP stack is a suite of software programs based on a TCP/IP model that provides services for Internet-based applications. The TCP/IP is the Internet communication protocol and the TCP/IP stack actually involves multiple protocols. The firmware founded upon TCP/IP stack is employed by the ColdFire MCF5223x family microcontrollers. The flow chart in Figure 4-26 summarises the flow of operations in relation to the TCP/IP stack.

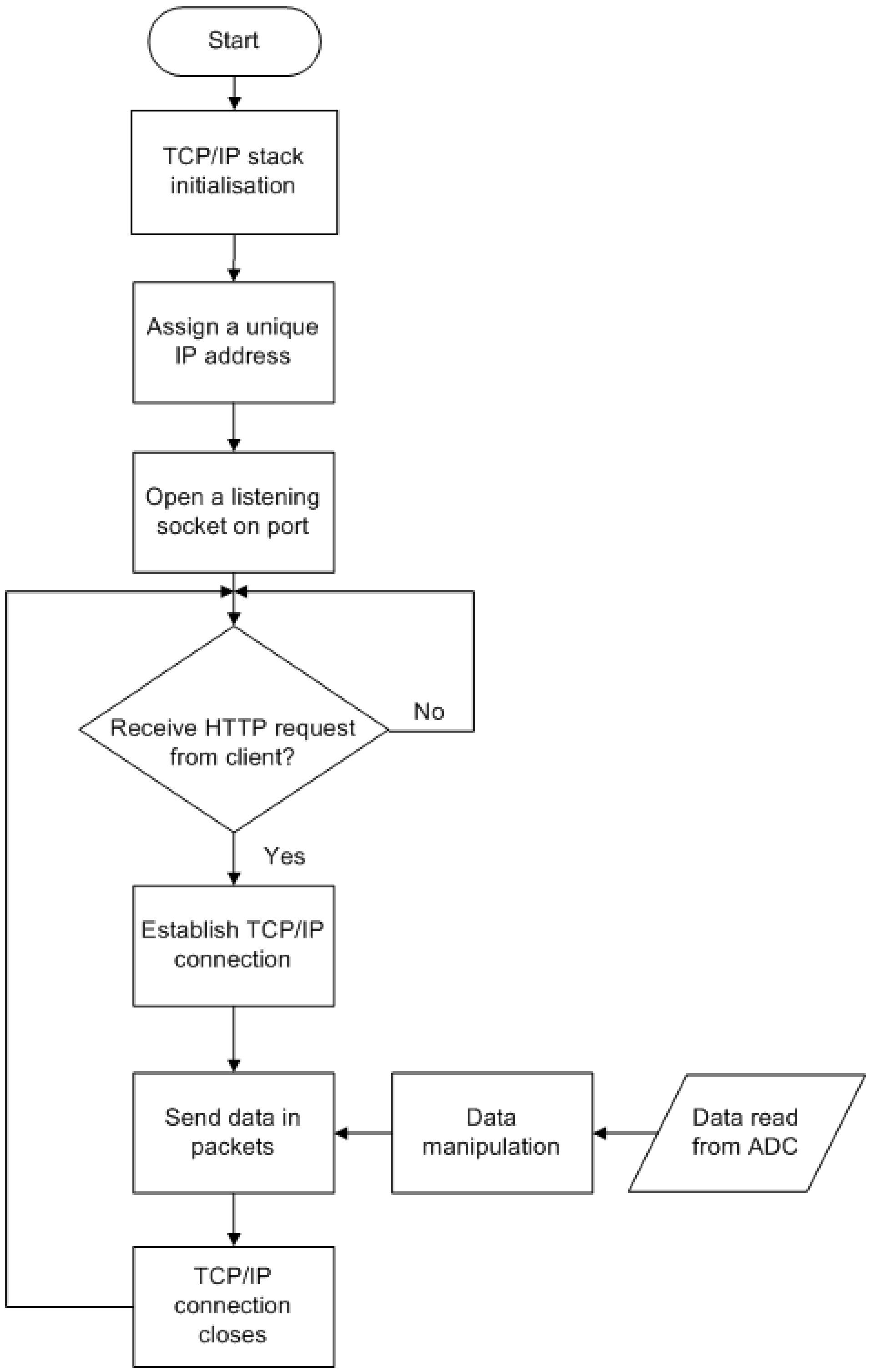


Figure 4-26 Flow chart of the TCP/IP stack with the firmware

During initialisation, the TCP/IP stack assigns a unique IP address and opens a listening socket on a port to await an HTTP request from client. While an HTTP request to this specified IP address is received, the stack establishes a TCP/IP connection between the server and the client through the socket. After the connection is accepted, the data which have been read from the ADC and been manipulated by the microcontroller are sent in packets to the client. This TCP/IP connection will be closed after the data transfer.

This stack is very robust and highly configurable, supporting most of the commonly used protocols. The table 4-3 shows the ColdFire TCP/IP stack.

Table 4-3 ColdFire TCP/IP Stack

<table>
<tr><td colspan="2">HTTP application</td></tr>
<tr><td colspan="2">TCP API</td></tr>
<tr><td colspan="2">TCP Layer</td></tr>
<tr><td colspan="2">IP layer</td></tr>
<tr><td colspan="2">FEC</td></tr>
<tr><td>Ethernet PHY</td><td>Hardware API</td></tr>
</table>

In the application of the research, the main features are listed in the following:

• HTTP

• TCP/UDP

• Serial-to-Ethernet

• Dynamic host configuration protocol (DHCP) or manual IP configuration

• Domain name server (DNS)

• User datagram protocol (UDP)

• Internet control messaging protocol (ICMP)

• IP

IP provides a mechanism to support multiple TCP connections, known as ports. Each IP connection is of a unique IP address and each network node on the Internet has an IP address (RFC2131, 1997). For example, every user has an IP address and every router has an IP address as well. With this system, the object's data could be acquiring by locating its unique IP address on the Internet.

TCP provides a negotiation system that establishes the connection by using an ACK and retry system. It offers a mechanism to guarantee data transmission from server to client (RFC2131, 1997).

DHCP defines a mechanism of acquiring network parameters like the IP address at runtime. It uses the UDP layer of the stack, and packets are transferred using the UDP layer and BOOTP ports. The stack employs a DHCP client to discover a DHCP server. The DHCP client discovers by means of broadcast address since the IP stack does not have an IP address yet. In the discovery packet, there is a unique transaction ID. When a listening DHCP server receives the transaction ID, it uses broadcast addressing to send an offer message containing the unique ID from the client and the suggested network parameters. When the client receives this server ID, sending a require packet back to the server indicating that it accepts the network parameters offered. After that, the server answers the client with the acknowledgement character (ACK) using its new IP address (RFC2131, 1997, RFC1035, 1987).

The DNS client communicates with the DNS Server. The aim of the DNS protocol is to translate the domain name into an IP addresses. DNS can use UDP or TCP with port 53. The DNS protocol is stateless and all the information is contained in a single message (RFC1035, 1987).

4.4.6 Transporting HTTP

The World Wide Web, abbreviated as WWW or known as the Web, is the most successful and dominant application/service running on the Internet. With a web browser, a range of information resources and services like text and images can be viewed on the Web across the Internet. This system exploits this capability to display the real-time biomedical signals either on a web page (the client) or LabVIEW software.

The success of the Web is due to the flexibility of HTTP. It is used to transfer data/files over the Internet. The HTTP client accomplishes the function of receiving and reading the data/files. A HTTP server is used for storing and transferring the data/files. HTTP defines the conversation between the server and the client over the TCP/IP Internet transport protocol. An HTTP server has been developed and integrated in the firmware, compatible with the stand-alone hardware board. The communication mechanism of HTTP transmission with TCP/IP stack is illustrated in Figure 4-27.

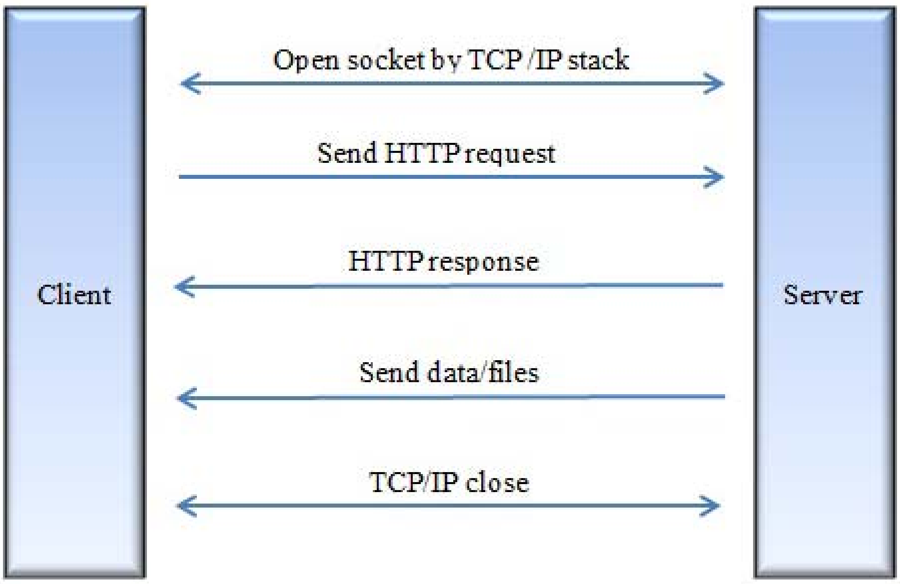


Figure 4-27 Communication mechanism of HTTP transmission with TCP/IP stack

The HTTP server listens for a connection on port 80 by default. In this application, the HTTP server applies the socket interface to create the listening socket on port 80 in order to avoid block from the router and fire wall.

Connection persistence and multiple session capability are the main features of the HTTP server. The compliant server supports multiple HTTP connections and includes a Flash File System (FFS) that supports ColdFire internal flash. It also supports long file names with subdirectories, which enables utilities for compressing run time and compile time downloadable images of multiple web pages.

In short, the designed stand-alone hardware board with the compatible firmware is ideal for applications requiring significant real-time operating for connectivity, data buffering and digital signal processing. The board has been successfully connected to the Internet and real-time biomedical signals can be viewed on the web page. By only using this module, the system can propagate the signals through the RJ45 cable. However, this mode is merely suitable for static applications, for example, patients who are bed bound or sedentary. Therefore, for ambulatory applications, a wireless solution should be included in the design.

4.5 Wireless adapter module

For ambulatory patients and mobile applications, a wireless communications system was added to the system. This was located between the front-end hardware and the access point to the Internet. The unit was required to have adequate range, compatibility with the Internet, low cost and be easy to install.

The wireless solution adopted was based on a wireless local area network which has interoperability among different network infrastructures. WLAN utilises the Wi-Fi protocol, which is a WLAN connectivity technology based on the IEEE802.11 standard. Wi-Fi is undergoing the same expansion in pace with the Internet, which makes it so successful. It is extensively adopted by most Internet interconnection networks so that Wi-Fi is particularly desirable for Internet-based system applications. A standard router should also serve as a wireless access point. At the other end, an Ethernet-to-wireless bridge switches the existing Ethernet connection of the designed module into the WLAN.

An Ethernet-to-wireless adapter which allows wired devices to connect to the WLAN should be employed here, explicitly, a LAN to WLAN bridge. Using the bridge, data packets can traverse two different type protocols: Ethernet (IEEE802.3) and Wi-Fi (IEEE802.11). The following functional block depicts the wireless design in Figure 4-28.

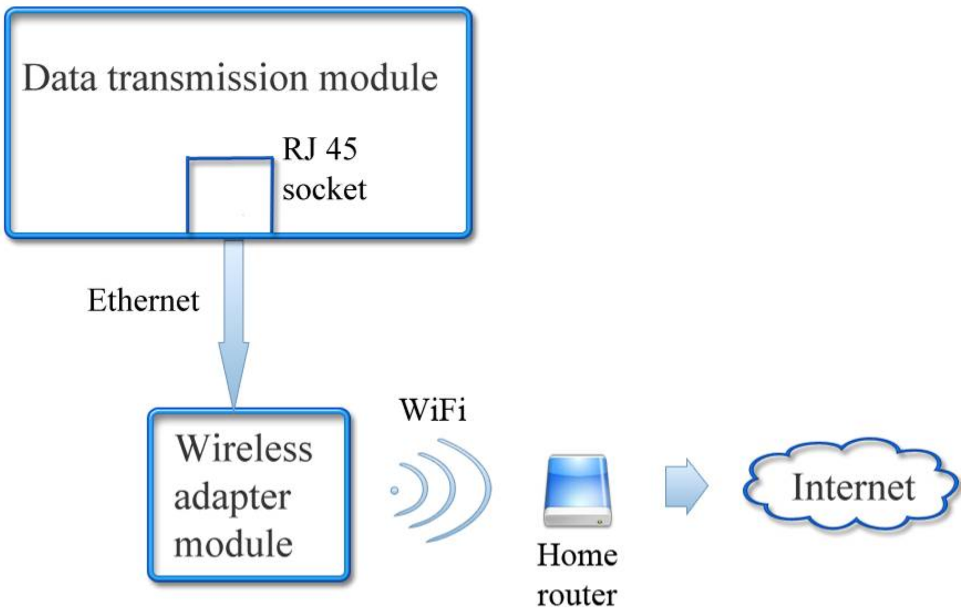


Figure 4-28 Functional block of wireless adapter module

An IEEE 802.11b/g wireless Ethernet adapter by TeleAdapt^{TM} for Wi-Fi environments was selected to accomplish the task of the Ethernet-to-wireless bridge. The adapter is designed to expand the existing wireless network infrastructure by offering a simple plug-and-play wired connection. Its RJ45 connector can easily plug into the designed stand-alone board. By this means, it bridges the sent data packets from the Ethernet to Wi-Fi towards the Internet. The wireless Ethernet adapter is shown in Figure 4-29.

Figure 4-29   A snapshot of the wireless Ethernet adapter

This adapter is supplied with an external AC mains power source 110 – 240 V with an output of 4 V DC, 1 A current. For ambulatory mode purpose, the AC main power supply circuitry is replaced with a battery DC supply. The IEEE 802.11 b/g standard operates at 2.4 GHz. The operating ranges are indicated by the manufacturer as 40 m indoors, 100 m semi-open and 457 m outdoor.

The wireless adapter links the RJ45 socket on the signal conditioning PCB board to communicate with the router wirelessly by Wi-Fi. The photograph in figure 4-30 shows the wireless adapter with the signal conditioning PCB board, the ECG leads and the data communication PCB board.

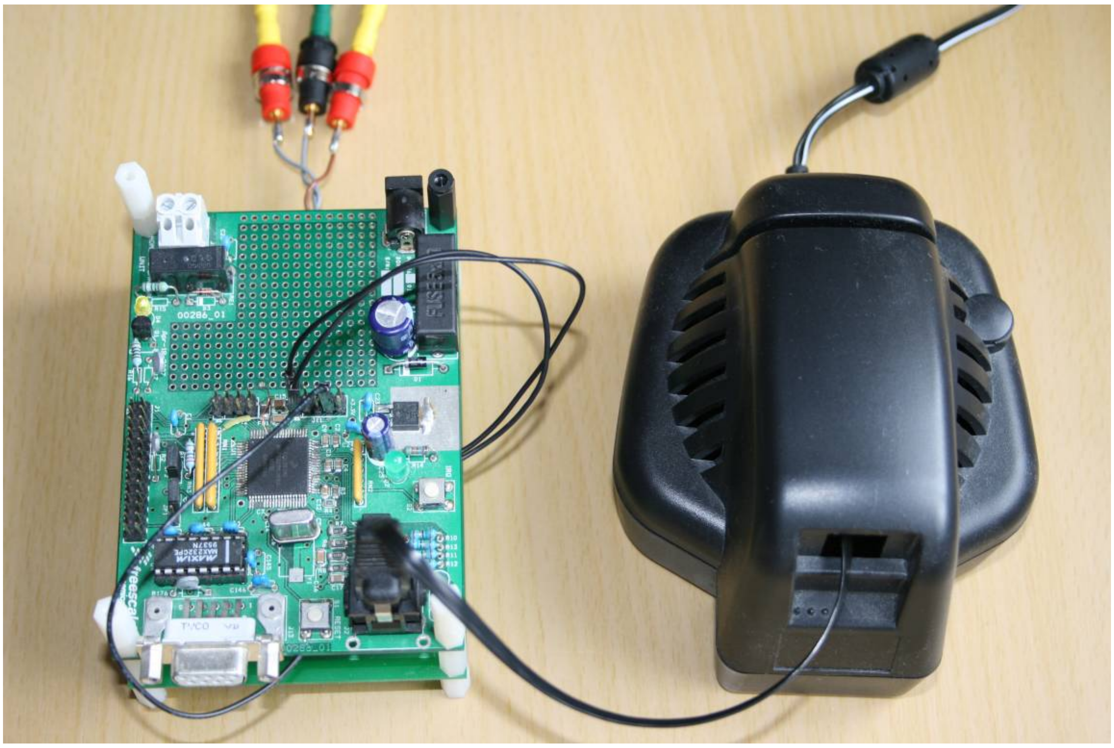


Figure 4-30 The wireless adapter with the signal conditioning PCB board, the ECG leads and the data transmission PCB board

4.6 Power supply module

The signal conditioning module, data transmission module and wireless adapter module need to be powered. This power supply module should address the task of supplying a steady voltage at the required current. The nature of the power source is dictated by the application: for static situations, a mains power source is the best solution. For mobile applications, a battery supply must be used. The functional block diagram is shown in Figure 4-31. Two different power supply sources will be considered.

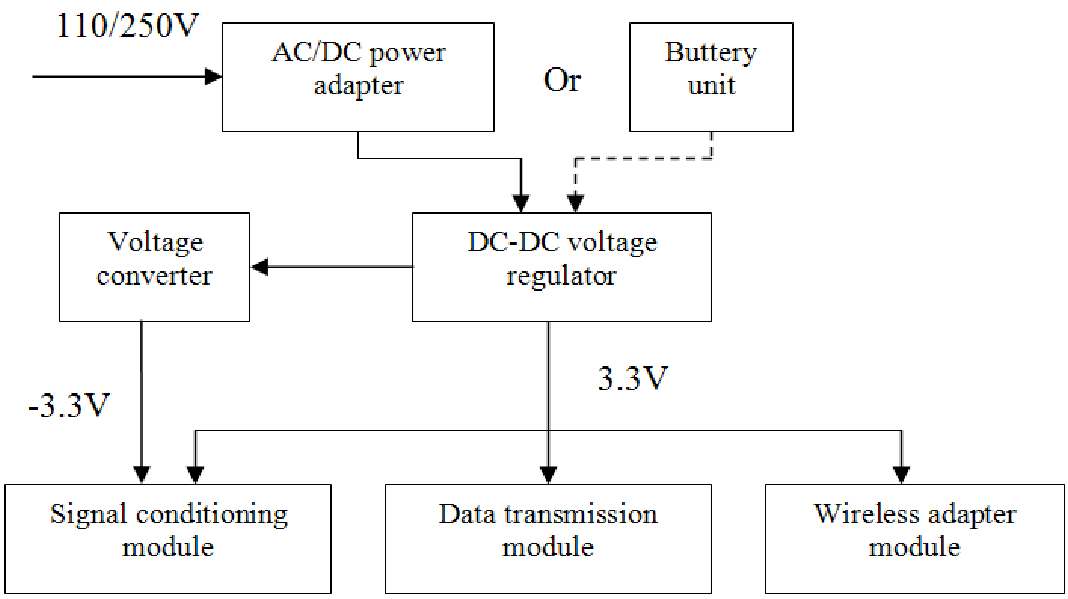


Figure 4-31 Functional block of power supply module in bedside/sleep and ambulatory modes

Many components in these modules are active ones, for example, amplifiers, microcontroller or RS232 transceiver. The power supplies to analogue conditioning module and to digital processing module should be isolated in order to avoid the noise interference each other. Table 4-4 summarises the power supply requirements of each component.

Table 4-4 A summary of components' power supply requirements

Module name	Component name	Power requirement
Signal conditioning module	Instrumentation amplifier	±3.3V
	Operational amplifier	+3.3V
Data transmission module	ColdFire Microcontroller	+3.3V
		+3.3V
	RS232 transceiver	+3.3V
Wireless adapter module		+3.3V

4.6.1 Static mode

If the power supply module is operating in static or bedside/sleep applications, a mains supply is preferable on account of considering stability and permanence. Hence an AC/DC power adapter with an output of 5-8 V DC at about 500 mA can fulfil the power supply for the circuits. Additionally, another power adapter supplied by AC mains power source delivers a maximum 4 V DC, rated at 1 A to the wireless transmission circuit. The voltage regulation circuit which is connected to the AC/DC power adapter is shown in Figure 4-32.

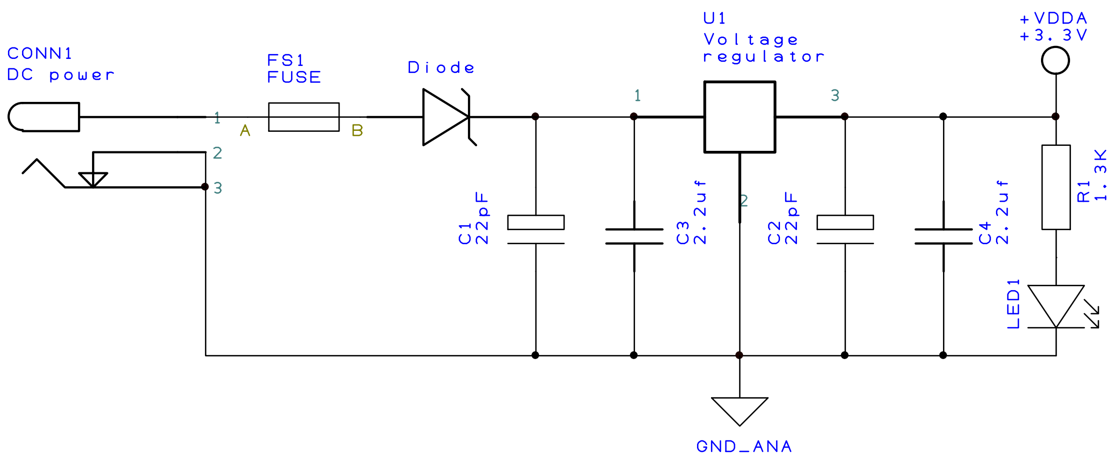


Figure 4-32 Circuit design scheme of power supply module provided by mains power

A rectifier converts the incoming AC voltage into DC. To protect the circuits and patient from over-current threat, an over-current protection fuse should be integrated into the circuit. Then a voltage regulator drops the input DC voltage down to constant 3.3V and provides a stable output.

4.6.2 Mobile mode

The designed system was required to have mobile capability. Hence, the miniaturised device necessitates the use of a portable and light-weight power supply unit. From this perspective, a battery supply was essential to provide the power to all circuits. The circuit layout is given in Figure 4-33.

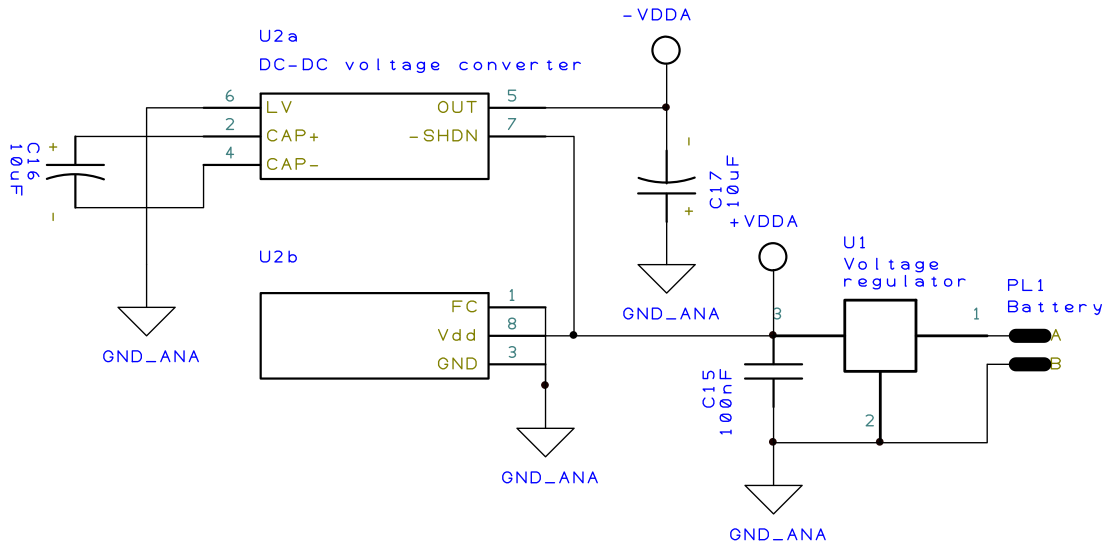


Figure 4-33 Circuit design scheme of power supply module provided by battery

The battery is connected to a very low drop DC-DC voltage regulator which produces a stable 3.3 V. From this, -3.3 V is generated by a voltage inverter to power the amplifiers. The capacity and rechargeable specifications of the battery are unimportant since this is a proof-of-principle system.

In the power supply circuit, due attention is given to mixed signal design. To isolate the low level analogue signals from a noisy digital environment, many data acquisition components have separate analogue and digital ground pins. In a similar manner, the digital 3.3 V and analogue 3.3 V supplies should have additional filtering by ferrite bead as the one is taken from the other.

4.7 Display and data storage module

4.7.1 Signal display

Once the physiological signals have been detected, processed and transferred via the Internet, they are ready for display and storage on the client computer (typically used by the clinician). The data graphing can be fulfilled either by the web browser or dedicated software.

In either the web browser or the LabVIEW scenario, a proprietary URL address which points to the patient's IP address is quoted. This execution directs the browser or the LabVIEW GUI to receive the signals from the transmission module in the biomedical system. The ECG data are plotted within the web page or the window of the GUI in real-time.

In many circumstances, authentication is required to restrict connectivity to personal devices, e.g. access to the patient's biomedical data, in order to protect confidential information. For this reason, this module employs encryption via a user name and password, restricting data access to authorised users.

4.7.2 Data storage

Practical and effective clinical software applications, required for monitoring and diagnosis, must not only receive and display biomedical signals, they must have the capability to store and retrieve the data for further analysis and diagnosis.

From the literature it is evident that most telemedicine systems provide the option of sending data to a centralised storage resource, usually network-attached servers. This solution results in an upsurge of the system cost and complexity. Besides, one clinician might not need full data in the database of the server but only some data collected from a limited number of patients. In addition, while an increasing number of patients will use the Internet based biomedical monitoring devices, the sufficient IP addresses provide less stress on the deficiency of the data base resource and the network bandwidth.

In the data storage software module, the storage function is implemented in the LabVIEW programme. The software has been designed to allow storage to a selected channel for saving the different categories of biomedical signals. Here is the programme layout of the data storage module in the LabVIEW development environment.

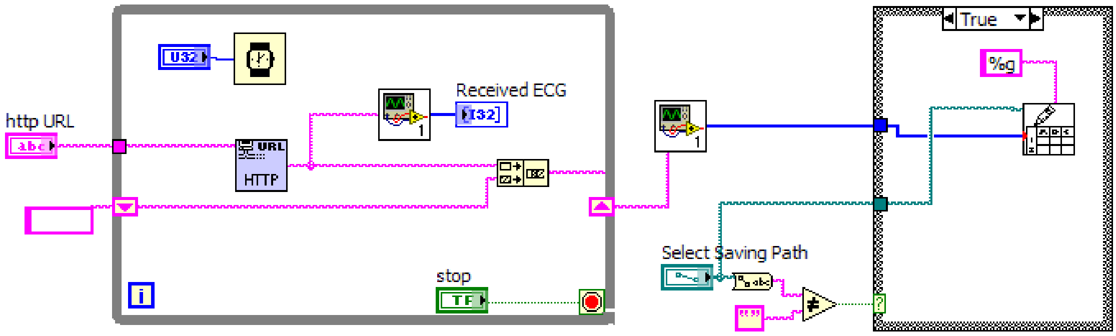


Figure 4-34 LabVIEW programme layout of display and data storage module

Again, the patient's proprietary URL address is designated. A reserved catalogue path should be indicated before saving the data. In addition, the LabVIEW programme can also plot the bio-medical signals.

The Savitzky-Golay digital filter is implemented in a sub-programme of the LabVIEW programme. It is applied exactly after the real-time data are received from the Internet. A second order of the polynomial is performed in the filter. This digital filter can smooth the dynamic curve of the ECG signal so as to diminish the level of noise with limited effect to the original data. The programme layout of the Savitzky-Golay digital filter is shown in Figure 4-35.

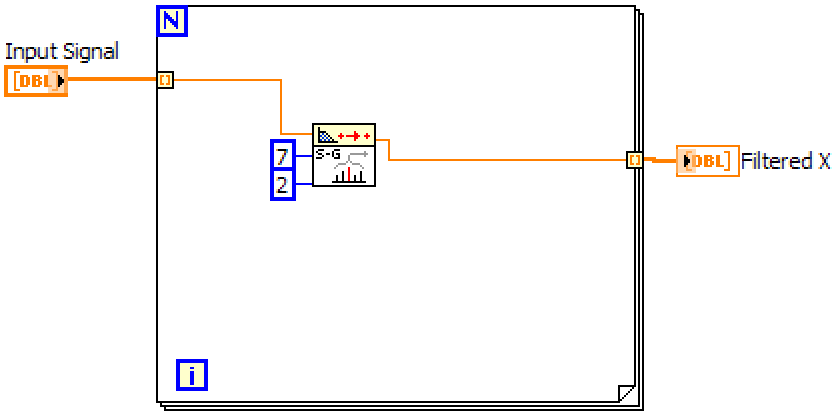


Figure 4-35 LabVIEW programme layout of the Savitzky-Golay digital filter

4.7.3 Data retrieval

The recorded data in the local hard disc can be retrieved as long as the storage progress has been completed. In the LabVIEW environment, a loading path should be selected to locate the required file which contains the data. The entire data can be recalled and re-displayed in the GUI. The programme layout of the LabVIEW programme is demonstrated in Figure 4-36. Alternatively, if the data is saved in an Excel document, it can be plotted directly by Excel facilities.

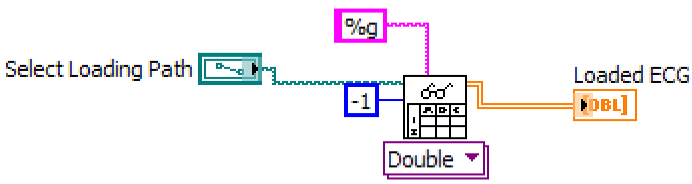


Figure 4-36 LabVIEW programme layout of data retrieval module

The combination of the Internet-based acquisition, display, storage and retrieval modules allow the user to view and store the data from any PC connected to the Internet. For further diagnosis, the dedicated LabVIEW software facilitates the storage of the data on the local hard disc of the client computer. By this means, it avoids the need for a centralized network with dedicated costly servers.

4.8 Summary

This chapter elaborates on the holistic design approach of the hardware and software platform for the Internet-based DSP system in the application of telemedicine. All modules in the system, including the multiple-sensor/channel module, signal conditioning module, data transmission module, wireless adapter module, power supply module and display/data storage module are described in detail design and function. New generation telemedicine system requires highly personalized and patient-centric monitoring, so each individual module has its own dedicated functionality to provide usability, reliability, portability and integration for this wireless Internet-based telemedicine system. The crux of the innovation is to design a truly embedded Internet-based telemedicine system rather than employing a computer, laptop, smart phone, PDA or other bulky equipments, as well as non-interoperable infrastructure/networks.

CHAPTER 5

Results and Discussion

5.1 Introduction

The prototype of the Internet-based real-time DSP system has been designed, constructed and tested in the research described here. This chapter first describes the evaluation of all sub-modules with respect to all applied biomedical signals. Second, an account is given of all experiments relating to biomedical signals acquisition (Ethics Approval was sought and provided). The physiological signals were obtained from the human bodies and manipulated by the system. The interpretation of these experimental results and the associated discussions are given. The performance of every module or sub-module is characterised also in this fashion. Moreover, some discussion of the results in relation to other comparative studies is presented in the last section.

5.2 System evaluation

This system acquires multiple biomedical signals in the study. The individual sub-modules manipulate real-time ECG signals, heart rate detection, respiration and body temperature respectively. In this section, evaluation of each sub-module is discussed.

5.2.1 Evaluation of the ECG and heart rate detection module

Quantitative evaluation of the system in relation to the ECG signals and heart rate detection was implemented using a Fluke Biomedical's MPS450 Multi-parameter Simulator. It provides physiological simulation of ECG signals with known heart rate and peak-to-peak amplitude parameters. The virtual ECG signals generated by the simulator were fed to the ECG conditioning module via the ECG leads. Figure 5-1 depicts the connection in the experiment. The ECG signals produced by the multi-parameter simulator are ideal, i.e. they are stable and noise-free.

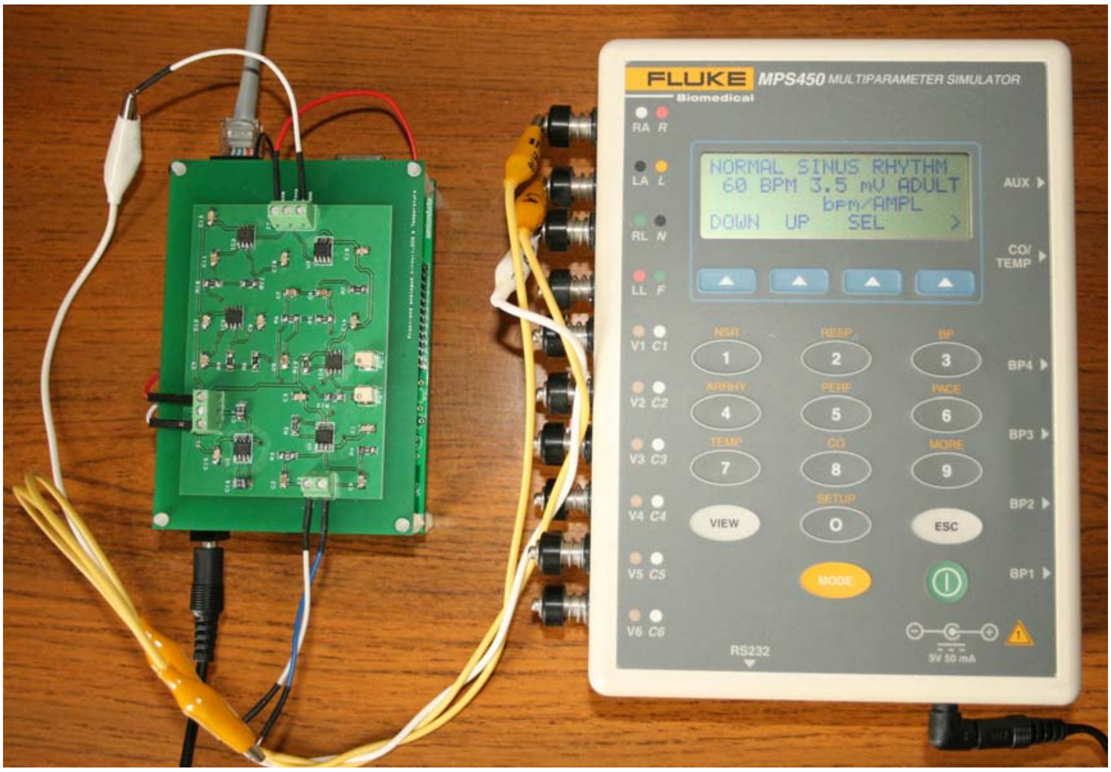


Figure 5-1 ECG signal simulation by MPS450 Multi-parameter Simulator

The heart rate and the peak-to-peak amplitude parameters of the ECG signals were set on the ECG simulator to 60 beats/minutes and 3.5 mV respectively. The simulated signals were manipulated by the front-end ECG signal conditioning module and fed to an oscilloscope, displayed in Figure 5-2.

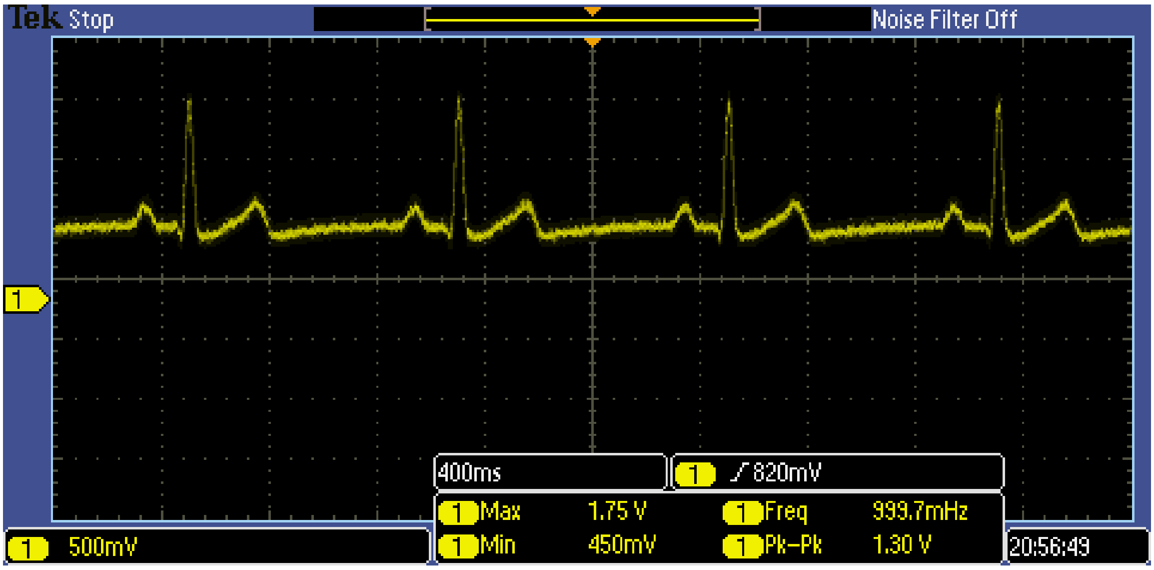


Figure 5-2 ECG signals manipulated by the ECG signal conditioning module

The processed signals with about 1.3V peak-to-peak amplitude are digitised and then transmitted through the Internet by the data transmission module. The gain of the ECG signals is governed by the signal conditioning module. The data display module, i.e., the LabVIEW program receives and reproduces the ECG signals in near real-time (delay is less than one second). In this LabVIEW program, the instantaneous heart rate is calculated based on the real-time ECG signals. The signal display and heart rate detection interface of the LabVIEW program is shown in Figure 5-3.

Figure 5-3 The signal display interface of the LabVIEW program via the Internet

From the user interface, it can be seen that the signals are shown with the same amplitudes as those prior to transmission. In this example, the heart rate was calculated by the software to be 60 BPM, shown in the ‘Heart rate’ display panel. This was the same value as generated by the ECG simulator. Hence, the simulated ECG signals with initialised parameters can be acquired, processed and transmitted via the Internet by the designed system. This system has the capability of accurately delivering the real-time ECG signals to the doctor/clinician end through the Internet.

Further analysis and evaluation of the ECG signals *in vivo* will be discussed in later sections.

5.2.2 Evaluation of respiration module

The performance linearity of the Pneumotrace II model 1132 transducer was examined next, respecting respiration measurement. To quantitatively evaluate the respiration module, the voltage outputs from the elastic piezoelectric sensor strap were measured using an oscilloscope. The experiment was conducted at normal room temperature. A mechanically accurate positioner and a vice were used to fasten two ends of the strap, which establish a coordinate system in order to measure the precise elongation of the strap. The following figure shows the layout of the experiment.

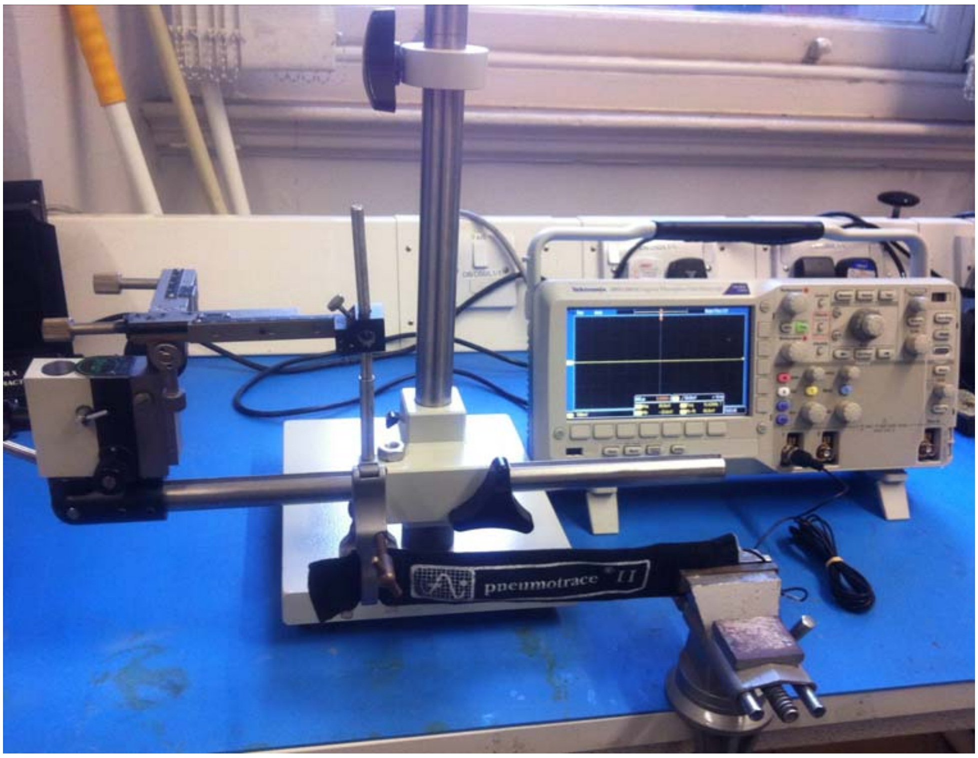


Figure 5-4 The respiration belt evaluation experiment

The sensor strap has a 300mm rest length. The length changes of the strap and the corresponding voltage outputs were recorded, listed in Table 5-1. After each stretching measurement, the strap should return to its original rest length without strain.

Table 5-1 The measured length changes of the strap and their corresponding voltage outputs

Test	Elongations (mm)	Voltage outputs (mV)
1	0	12
2	5	61
3	10	194
4	15	316
5	20	450
6	25	581

The transfer function is shown in Figure 5-5.

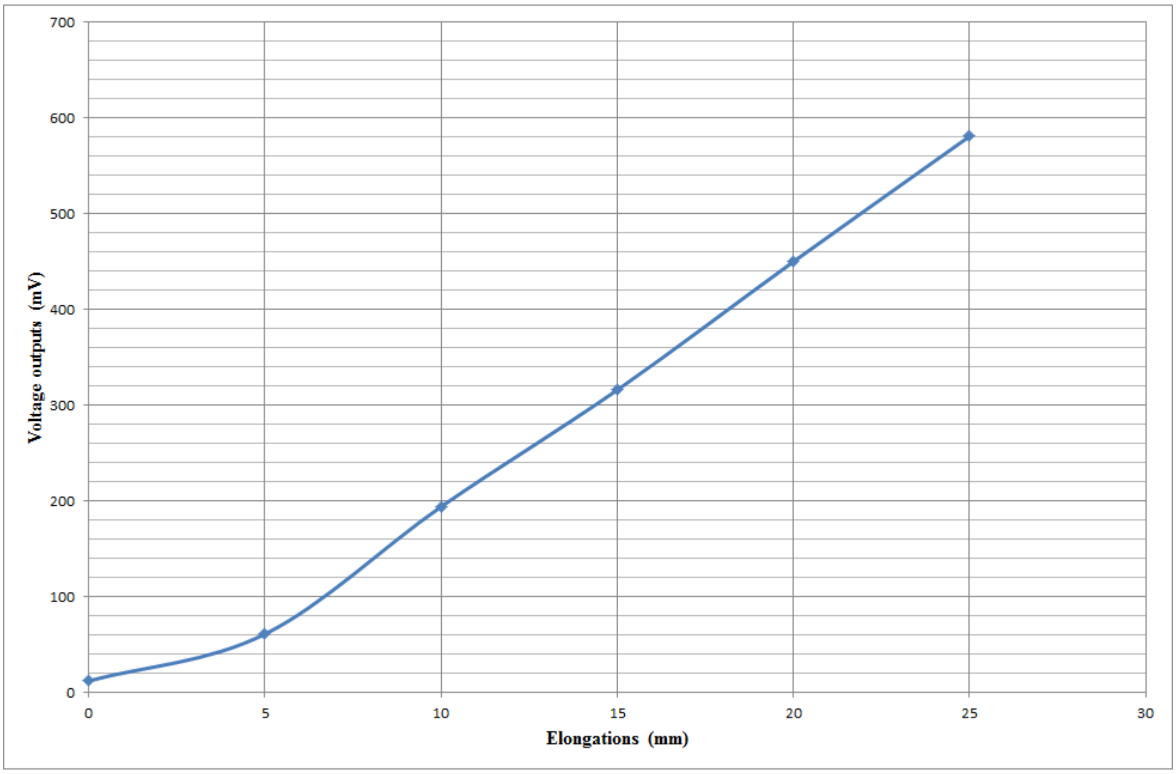


Figure 5-5 The variation curve of the output measurement against the elongations

The x-axis represents the length change of the sensor strap and the y-axis represents the measured output voltage.

The evaluation experiment was repeated a further five times; the results are recorded in APPENDIX B and plotted in Figure 5-6.

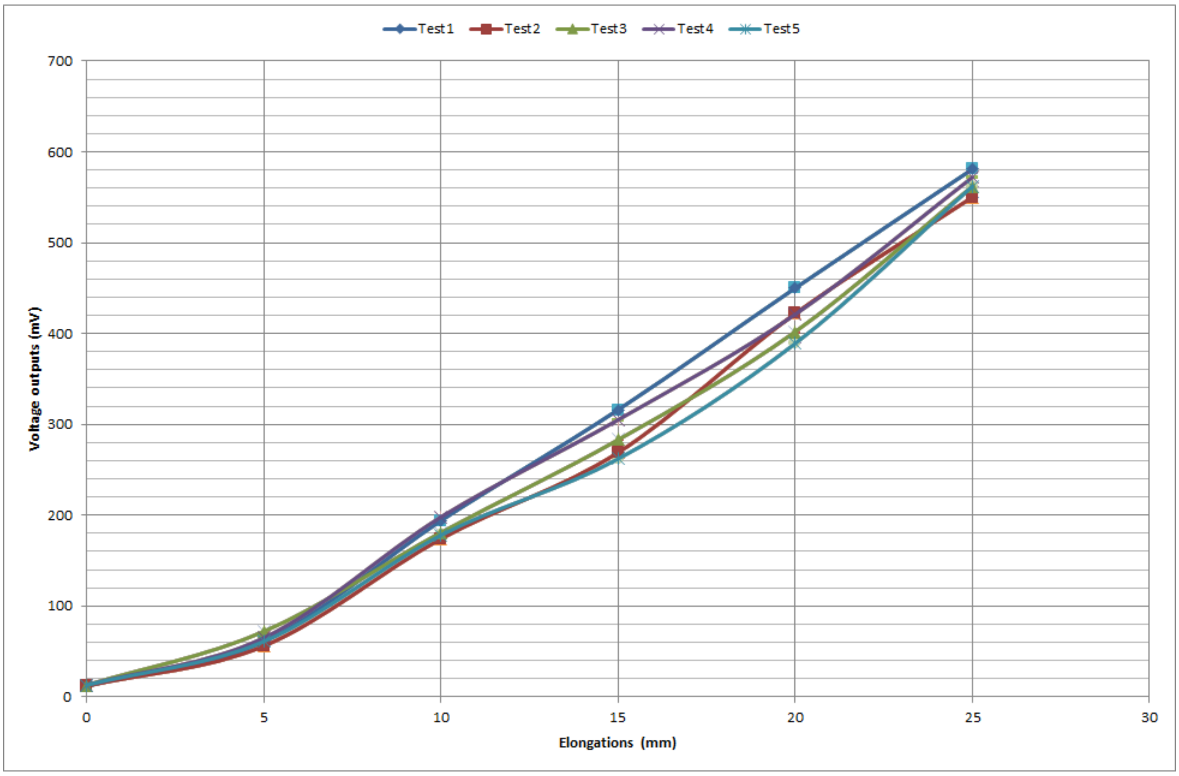


Figure 5-6   A set of readings for the respiration transducer evaluation

Although the curve is initially non-linear, about 7 mm straightens and was therefore deemed suitable measuring respiration.

5.2.3 Measurement of body temperature

The linearity and response time of the body temperature sensor was quantitatively evaluated next. The temperature sensor was submerged in warm water within a container, and as the water cooled, readings were taken from the sensor and a thermometer also immersed in the water. The temperature ranged from approximately 45 ˚C down to 30 ˚C. The results are shown in Table 5-2 and the variation curve of the sensor measurement against the thermometer reading is plotted form in Figure 5-7.

Table 5-2 The temperature readings by the temperature sensor and the thermometer

Test	Reading by the sensor (mV)	Reading by the sensor (°C)	Reading by the thermometer (°C)
1	968	47.5	47.8
2	934	43.6	45.4
3	920	42.0	43.2
4	900	40.3	41.6
5	882	38.6	39.6
6	866	37.0	38.2
7	851	35.9	36.6
8	840	34.7	35.4
9	826	33.3	34.2
10	810	31.6	31.8

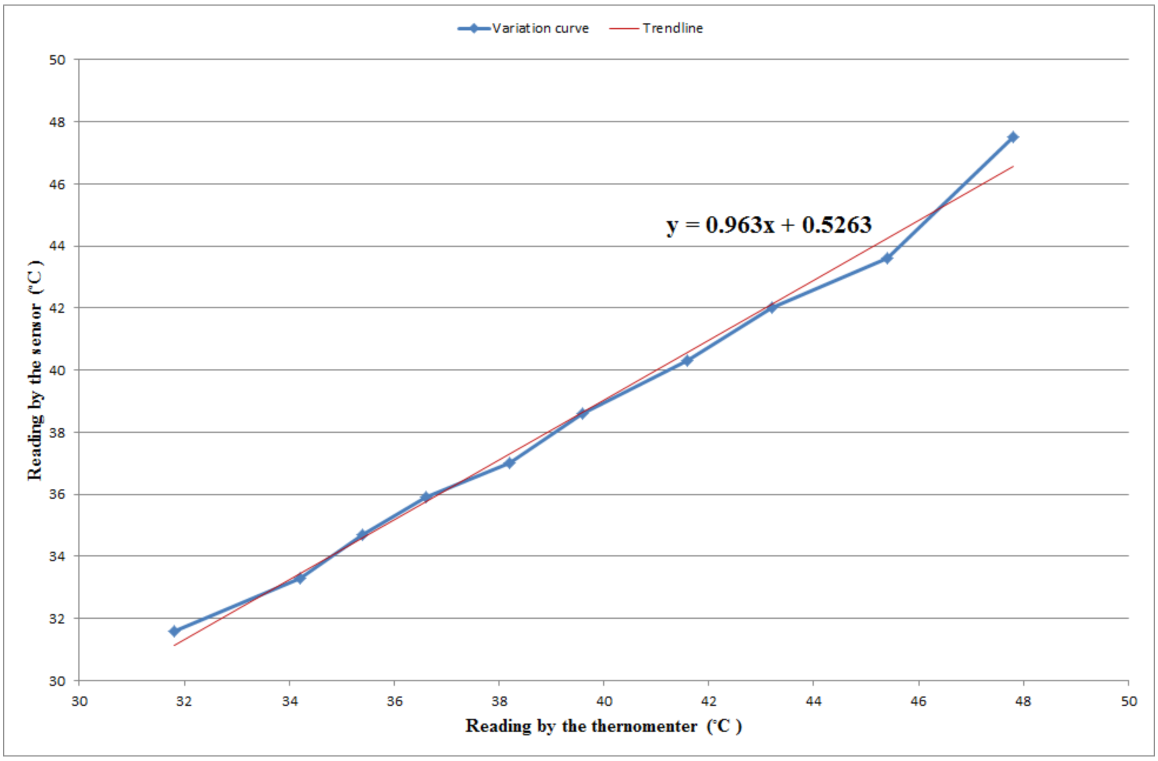


Figure 5-7 The variation curve of the sensor measurement against the thermometer measurement

In Figure 5-7, the x-axis denotes the temperature readings by the thermometer and the y-axis represents the corresponding measurements from the sensor. The blue line represents the variation curve of the sensor measurement against the thermometer readings. The red one refers to an estimated trend line toward the curve. Accordingly, a calibration equation of the trend line can be given in order to adjust the sensor measurement to accurate value.

$$y = 0.963x + 0.5263 \tag{5-1}$$

Where x represents the readings by the thermometer and y records the corresponding sensor measurements. Accordingly, the acquired temperature values by the sensor can be calibrated by Equation 5-1 to gain more accurate measurement. A set of repeated tests are taken and plotted in Figure 5-8. The results are listed in APPENDIX C.

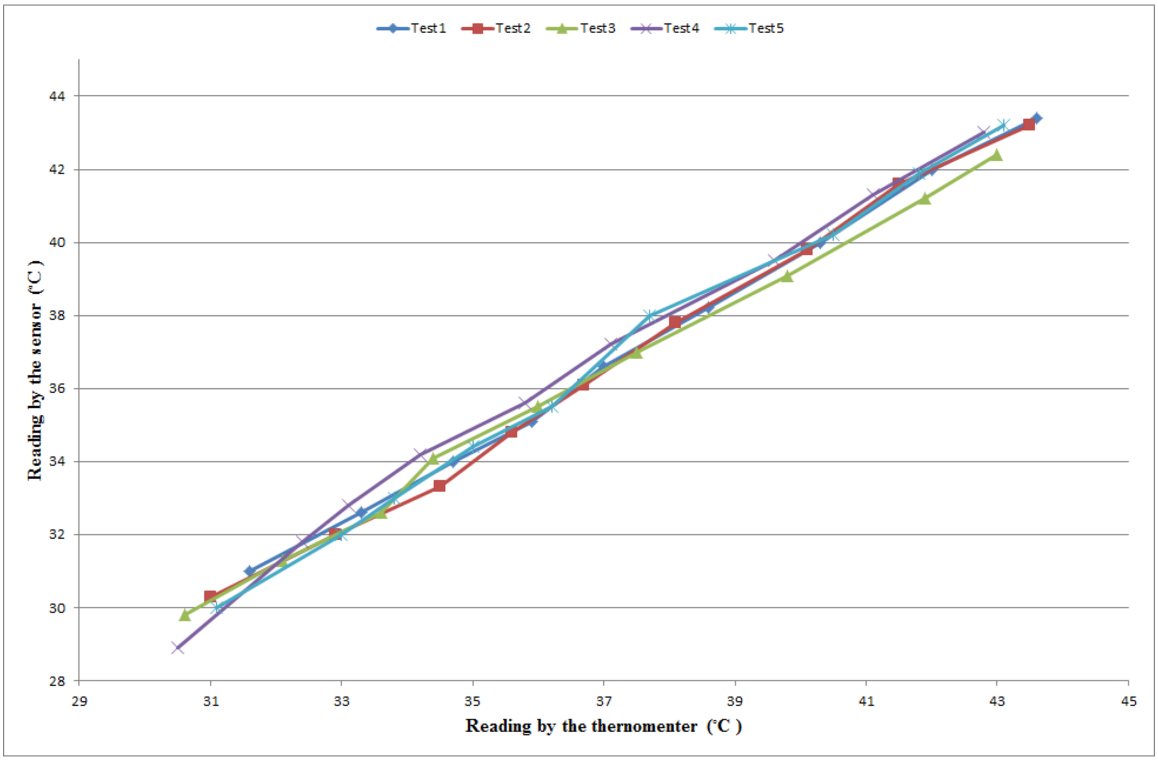


Figure 5-8 A group of repeated measurements for the temperature sensor evaluation

The relative deviations between the sensor measurements and the thermometer readings are illustrated by the variation curves, shown in Figure 5-8. The main deviation may be introduced by the delay of the responding time whilst the temperature of the water is still decreasing during measurement, or may occur due to the sensor's accuracy.

To determine the response time of the sensor, it was immersed in a container of water heated to 35 $^{\circ}$C, and the output was monitored. Initially, the sensor in air, prior to immersion, was at a temperature of 16 $^{\circ}$C. As Table 5-3 and Figure 5-9 show, approximately 20 seconds elapsed before the maximum output reading appears.

Table 5-3 The measurements of the sensor response time

Elapsed time (s)	Measured Voltage (mV)	Estimated Temperature ($^{\circ}$C)
0	662	16.2
5	778	27.8
10	803	30.3
15	822	32.2
20	837	33.7
25	841	34.1

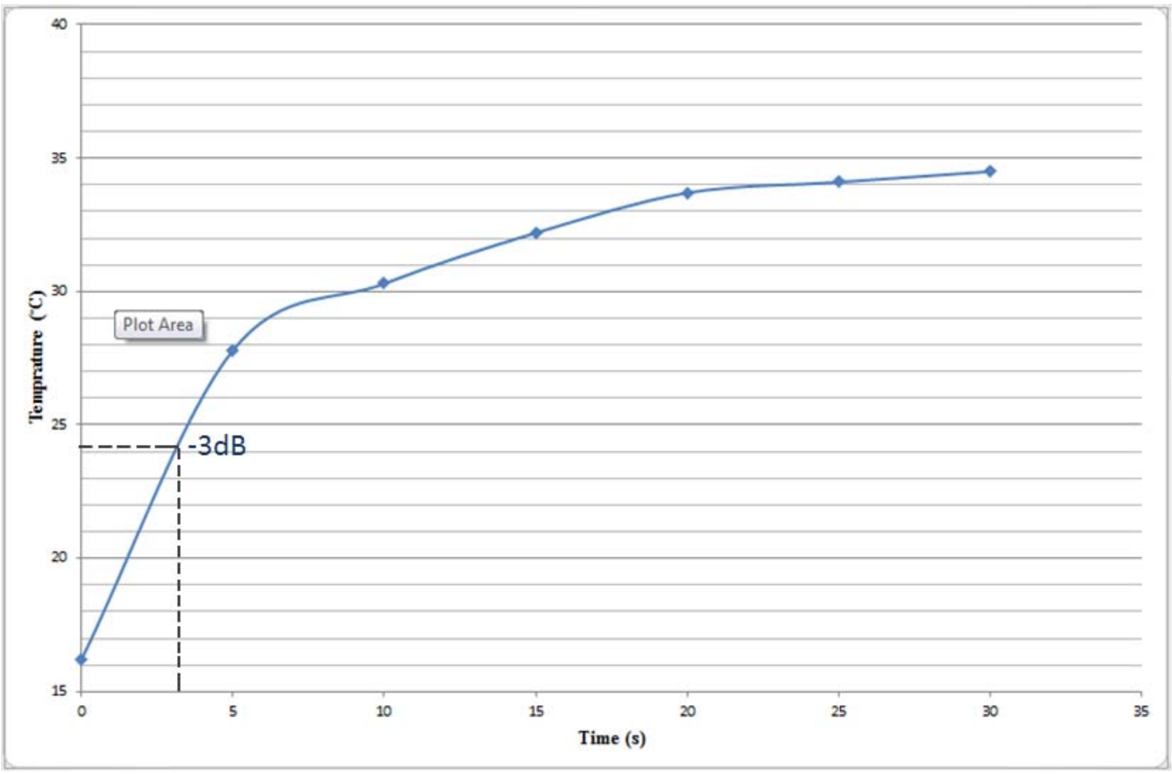


Figure 5-9 The variation curve of the sensor response time

From the figure above, it is observable that the temperature sensor takes approximate 4 seconds to reach a stable output reading (the -3dB point, measured from 0 ˚C, is introduced as the start point-in-time to collect the valued data); this was considered the response time. When the temperature sensor is applied *in vivo*, the readings should be taken after at least 4 seconds to ensure accuracy. Generally, fluctuations in body temperature occur over periods of several minutes (not least due to thermal inertia), so the sensor was deemed acceptable for the purpose of body temperature measurement.

5.2.4 Power consumption

When the system is operating in the mobile mode and is supplied by battery, the power consumption is an essential parameter of the performance. The power consumption of the signal conditioning module, data transmission module and wireless adapter module were tested using the mains adaptor.

Each individual PCB was characterised for power consumption by connecting a multimeter, configured to measure current, in series with the DC supply. Readings were taken once the supply voltage was stabilized. After powering the board, the stable reading of the multimeter gave the current drawn by the board. The product of the current reading with the supply voltage yielded the power consumption of this board. The measured results are listed in Table 5-4.

Table 5-4 Power consumption of each module

Module name	Voltage	Current	Power consumption
Signal conditioning module	3.3 V	6 mA	About 20 mW
Data transmission module	5 V	341 mA	About 1705 mW
Wireless adapter module	3.3 V	480 mA	About 1584 mW
Total			About 3309 mW

The front-end active units of the system consume about 3309 mW in total, which can clarify the power consumption of the portable device supplied by the battery.

5.3 Initial signal conditioning results

With the purpose of determining the inherent characteristics of the system, all experiments were performed *in vivo* respecting all chosen biomedical signals. Here, the Lead II ECG signal is discussed first. The effects of amplification, various filtering and level shifting functionalities were tested and verified. Dual-channel examination with the Lead I and Lead II ECG signals is given subsequently. In addition, the respiration and body temperature signals were also tested *in vivo*.

5.3.1 ECG electrodes and leads

The Ag/AgCl electrode with electrolytic gel-adhesive is employed to sense the ECG signals. Three electrodes are deployed on the torso to obtain the ECG signals Lead I and Lead II. They are 'LL' place on the pro-thorax above the left hip, 'RA' set below the right clavicle and 'LA' positioned below the left clavicle on the thorax. The cables connected to these electrodes are twisted around each other along their lengths, and conduct the ECG signals to the front-end signal conditioning module. The illustration shows the arrangement in Figure 5-10.

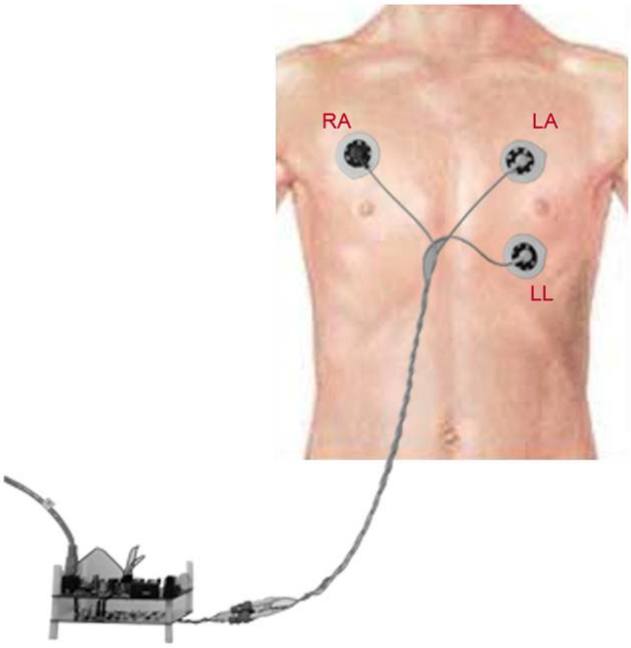


Figure 5-10 Illustration of the electrodes and leads arrangement

5.3.2 Performance of amplification

Through the lead cables, sensed analogue ECG signals are fed to the signal conditioning unit. The AD620 instrumentation amplifier performs signal amplification with a high CMRR. According to equation (3-12), the theoretical value of R_G is 49.45 Ω. In practice, the amplifier with a 47.5 Ω resister yields a gain of 1041 in accordance with the equation (3-11). To validate it, a 100 Hz sine wave signal with 5 mV peak-to-peak amplitude produced by a function generator, was fed to the AD620. The magnitude of the amplified output signal was measured by the oscilloscope, a trace from which is shown in Figure 5-11.

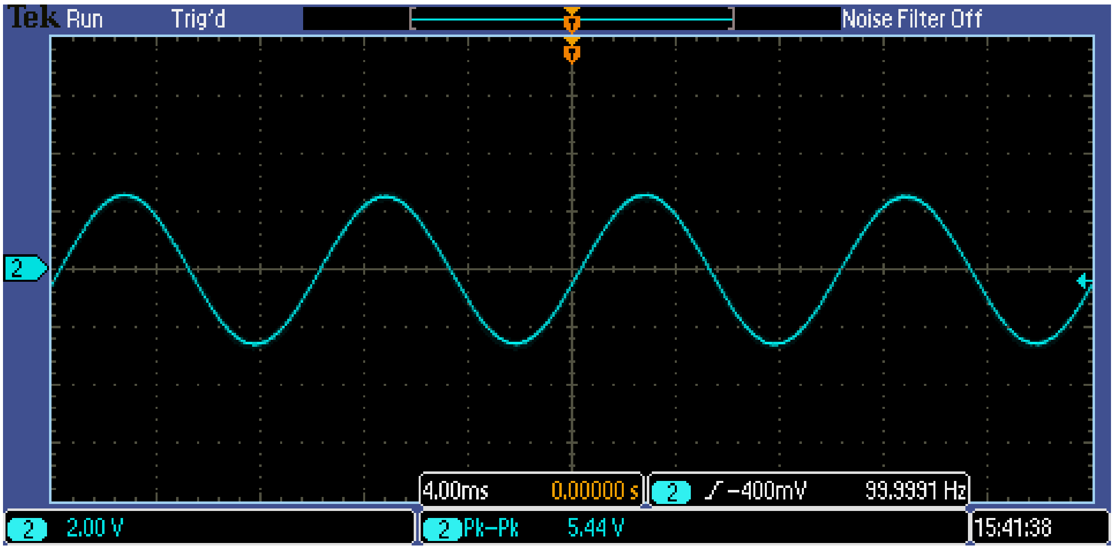


Figure 5-11 The measured peak-to-peak voltage of the amplified signal on the oscilloscope

The peak-to-peak value was 5.44 V as measured. Therefore, the actual gain given is:

$$\frac{5.44V}{5mV} = 1088 \tag{5-2}$$

The error is approximately:

$$\frac{1088 - 1041}{1041} \times 100\% = 4.51\% \tag{5-3}$$

Although the actual gain exceeds the designed value, as the amplitude range of the ECG signals still falls within the range of the ADC input, the error can be accepted by the system.

As an important specification of the instrumentation amplifier, the CMRR indicates how the common-mode voltage signal (considered as noise in this application) is rejected or attenuated in the output measurement. To determine the CMRR performance of the AD620 amplifier, a test circuit was designed as shown in Figure 5-12.

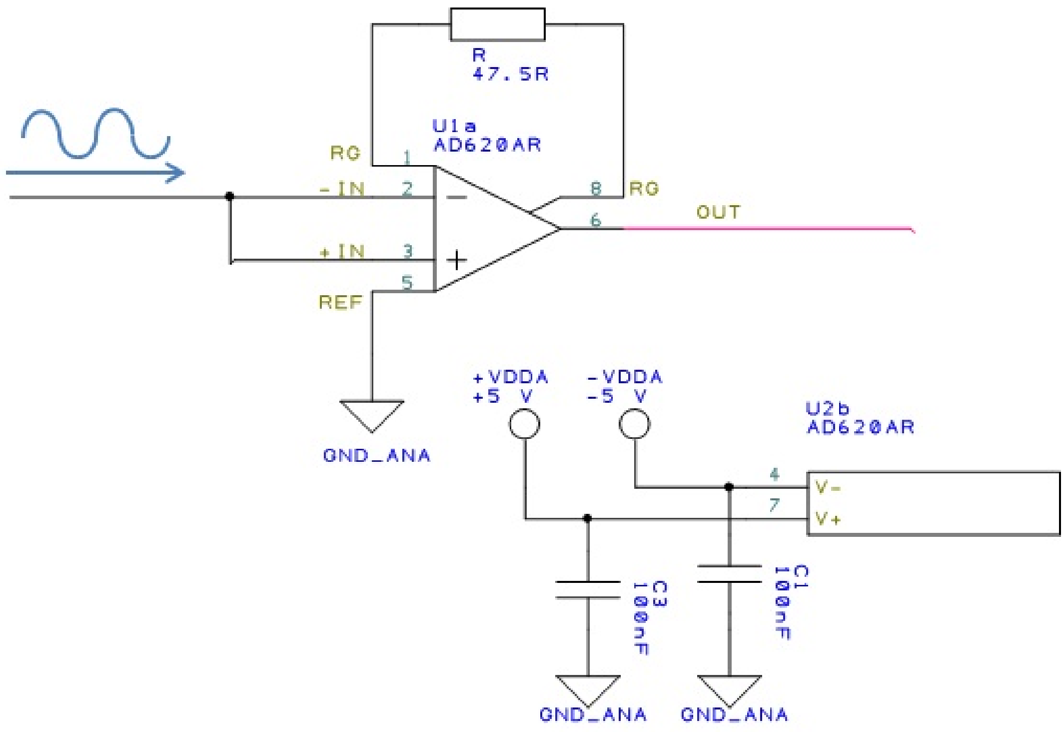
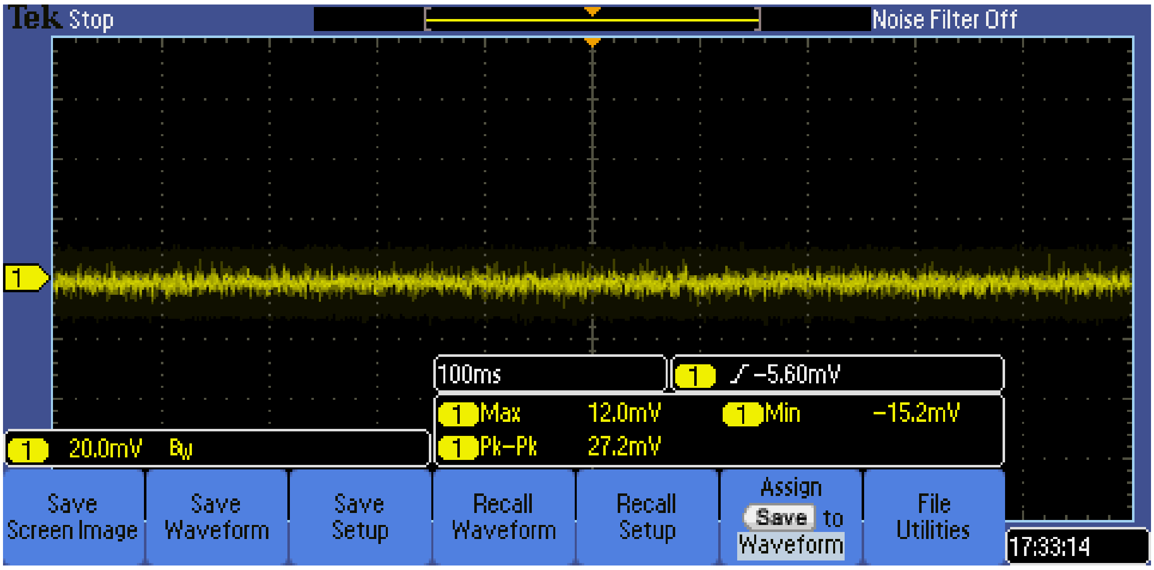
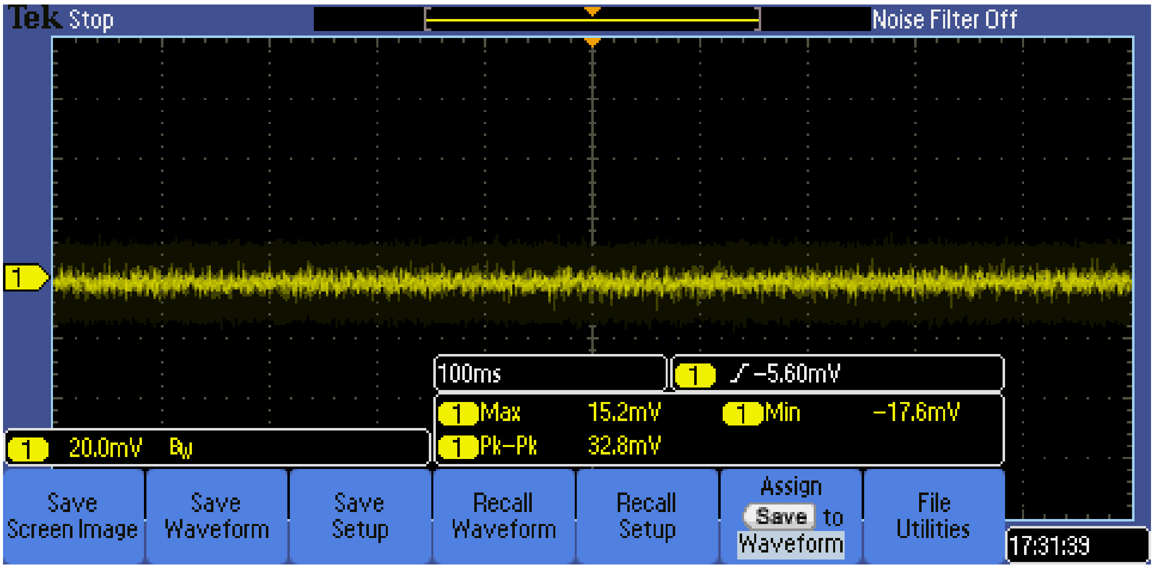


Figure 5-12 Circuit layout for CMRR performance evaluation

The power rails of the AD620 amplifier were connected to ±5 V DC in the test, yielding a permissible input signal voltage range of between –3.1 V and +3.8 V (Analog Devices, 2004). The 47.5 Ω resistor was also required to produce a gain of 1041. Two sine waves produced by a function generator, at 0.5 Hz and 100 Hz, both with peak-to-peak amplitudes of ±3 V were fed to the two inputs of the instrumentation amplifier simultaneously. A peak-to-peak amplitude of ±3 V was adopted in order to establish the full common-mode input voltage immunity. CMRR analysis was performed using a frequency sweep extending between 0.5 Hz to 100 Hz. The output voltage magnitude responses were measured and displayed on the oscilloscope as shown in Figure 5-13.



(a)



(b)

Figure 5-13 The magnitude of the output voltage response by applying the common-mode signal at (a) 0.5 Hz, (b) 100 Hz

The corresponding output offset voltage error produced by the amplifier was relatively low (about 30 mV which includes the ADC noise of the oscilloscope). It was not possible to see or measure breakthrough of the AC signal on the output. Therefore, this evaluation confirmed that the CMRR performance of the AD620 amplifier was suitable for use in this biomedical signal application.

5.3.3 Result and performance of the high-pass filter

The analogue first order RC high-pass filter is utilised to inhibit baseline drifting and probable 300 mV voltage bias to avoid amplifier output saturation, as discussed in section 4.3.2. The stop frequency of this filter is stipulated at 0.5 Hz. The theoretical value of the cut-off frequency based on the components in the RC circuitry is:

$$F = \frac{1}{2\pi RC} = \frac{1}{2 \times \pi \times 68k\Omega \times 4.7\mu F} = 0.498Hz \qquad (5\text{-}4)$$

Where R is $68k\Omega$ and C is $4.7\mu F$.

The performance of the high-pass filter may be portrayed by the frequency response of the RC circuit. To do so, the function generator produces a series of sine waves at 2 V peak-to-peak amplitudes in escalating frequency harmonics. This set of sine waves are fed to the input of the filter and its outputs are measured by the oscilloscope.

The normalised amplitude values may be plotted in a frequency spectrum coordinate axis. The frequency response of the high-pass filter can be charted as a curve in Figure 5-14, in which the vertical coordinates label the normalised amplitudes.

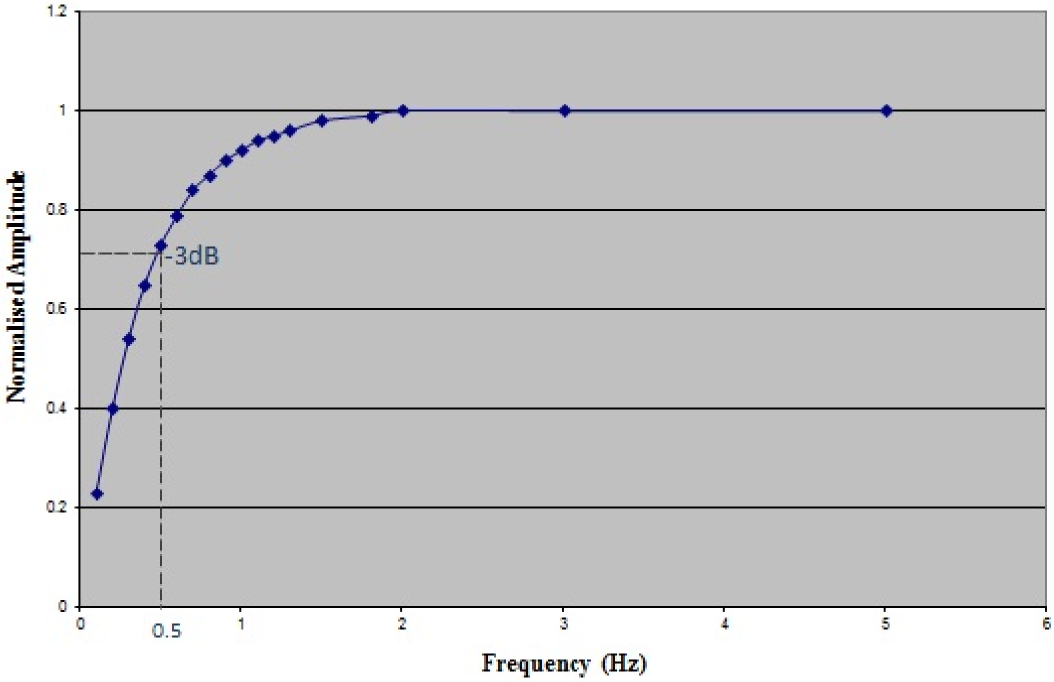


Figure 5-14   Frequency response of the high-pass filter

The combined effects of both the high-pass filter and the amplifier applied to the raw ECG signals are given as an oscilloscope trace in Figure 5-15.

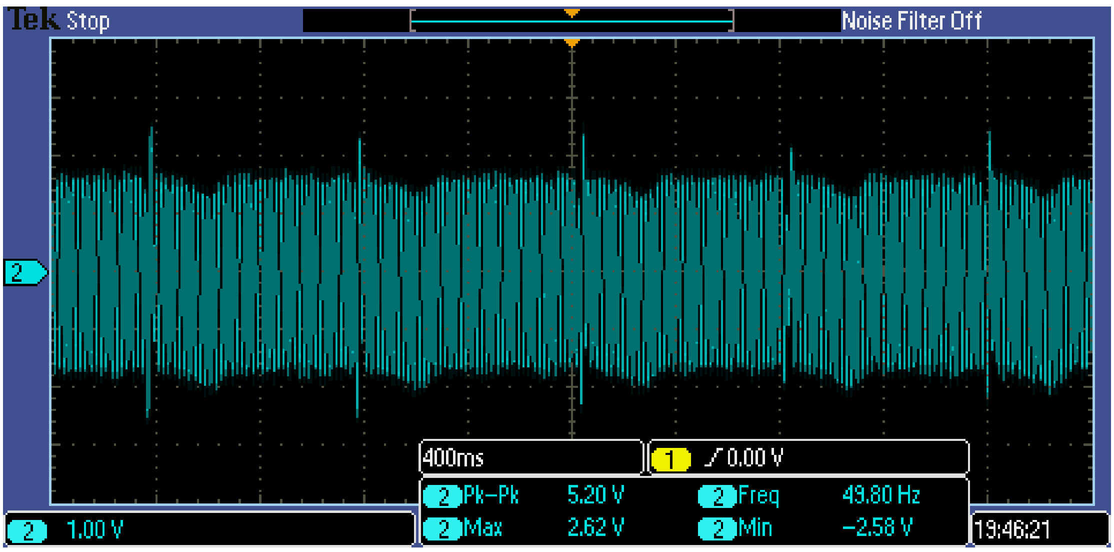


Figure 5-15   Result of the high-pass filter and the amplifier

From the measurement reading in the figure, the baseline of the signal is close to zero and the signal's amplitude range is –2.58 V to 2.62 V. Because the subsequent filtering application employs single supply power operation, the full signals are required to be shifted to a unipolar range.

In the figure, it is apparent that visible noise exists in the signal. To identify it, further examination was done using the cursor measurement feature of the oscilloscope, as shown in Figure 5-16.

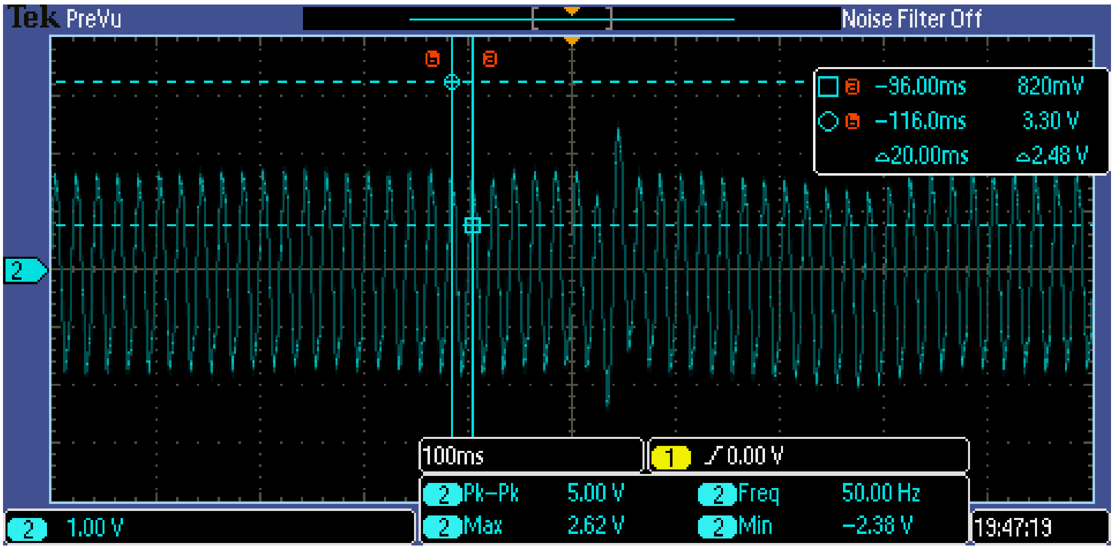


Figure 5-16 Measurement result by the cursors in the high-pass filter and the amplifier

The frequency of the noise was ascertained using the measurement feature of the oscilloscope. The visible noise represents a period of 20 ms, that is, contamination in 50 Hz frequency. Alternatively, if the noise is observed by using Fast Fourier Transform (FFT) algorithm, the frequency spectrum of it can be indicated in Figure 5-17. It is noticeable that the main power of the noise distribution locates at the centre of 50 Hz.

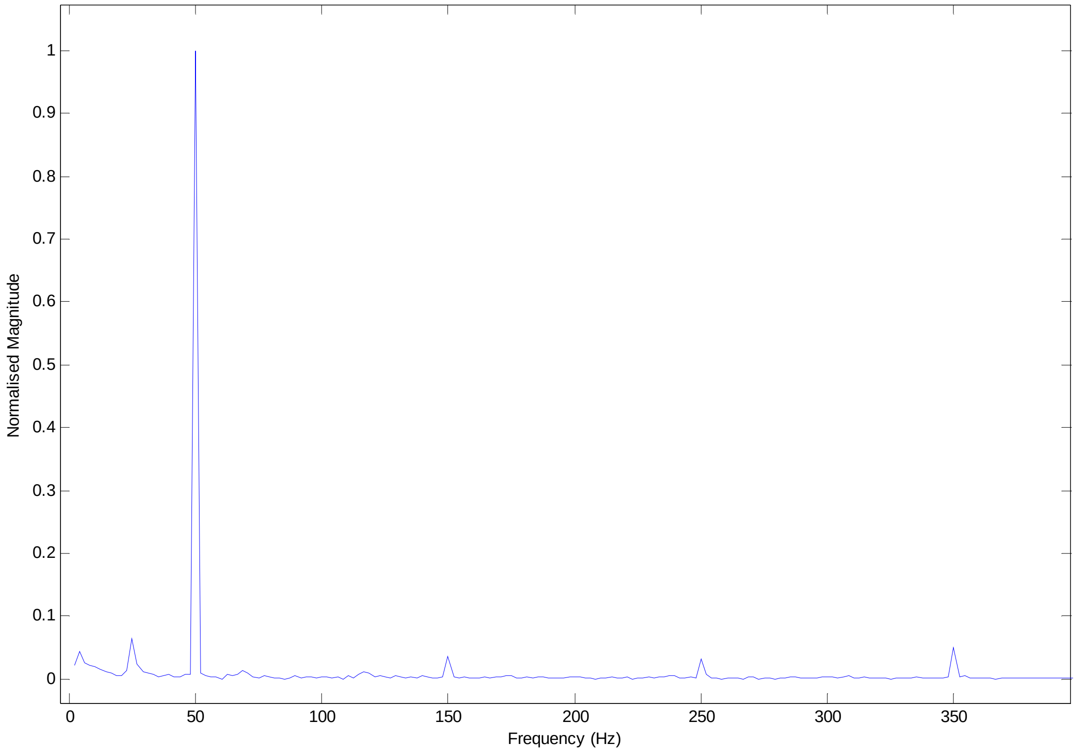


Figure 5-17   Frequency spectrum of the noise

5.3.4 Result and performance of the notch filter

Frequency analysis confirms that the ECG signals are still obscured by various noises sources, mainly 50 Hz interference, before being processed by the notch filter. Figure 5-18 reveals the effect of the noise on the ECG signals in the frequency domain.

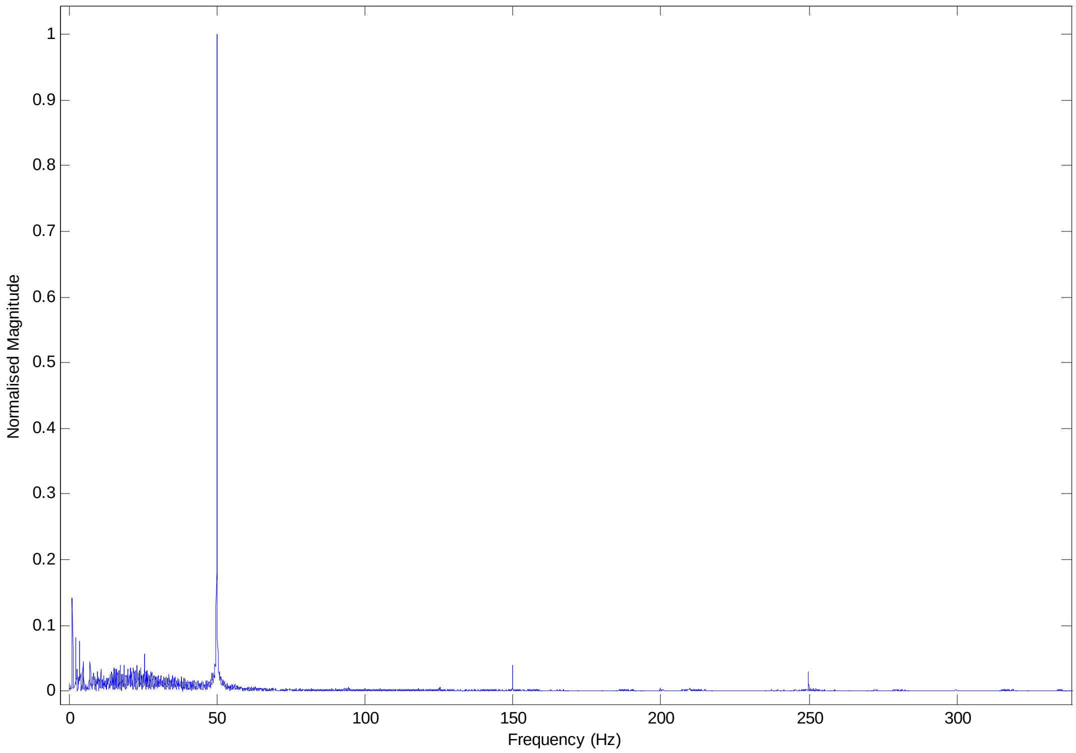


Figure 5-18 Frequency spectrum of the ECG signal before processed by notch filter

In order to cope with 50 Hz interference, the analogue twin-T notch filter with a centre frequency 50 Hz was employed. According to the computational formula of a twin-T notch filter, the estimated central frequency was:

$$F = \frac{1}{2\pi RC} = \frac{1}{2 \times \pi \times 3.3k\Omega \times 1\mu F} = 48.2Hz \tag{5-5}$$

Consistent with the circuit design scheme in Figure 4-15, R is $3.3k\Omega$ and C is $1\mu F$.

To ascertain the performance of the notch filter, the following experiment was conducted to determine the frequency response. The filter was probed using a frequency-swept sine wave, ranging from 1 to 700 Hz. The filter response was measured using an oscilloscope. The peak-to-peak amplitudes of the outputs at various different frequencies were recorded and plotted.

If the normalised amplitude values are plotted in the frequency domain, the frequency response of the notch filter can be revealed in Figure 5-19. In the figure, the vertical coordinates dictate the normalised amplitudes.

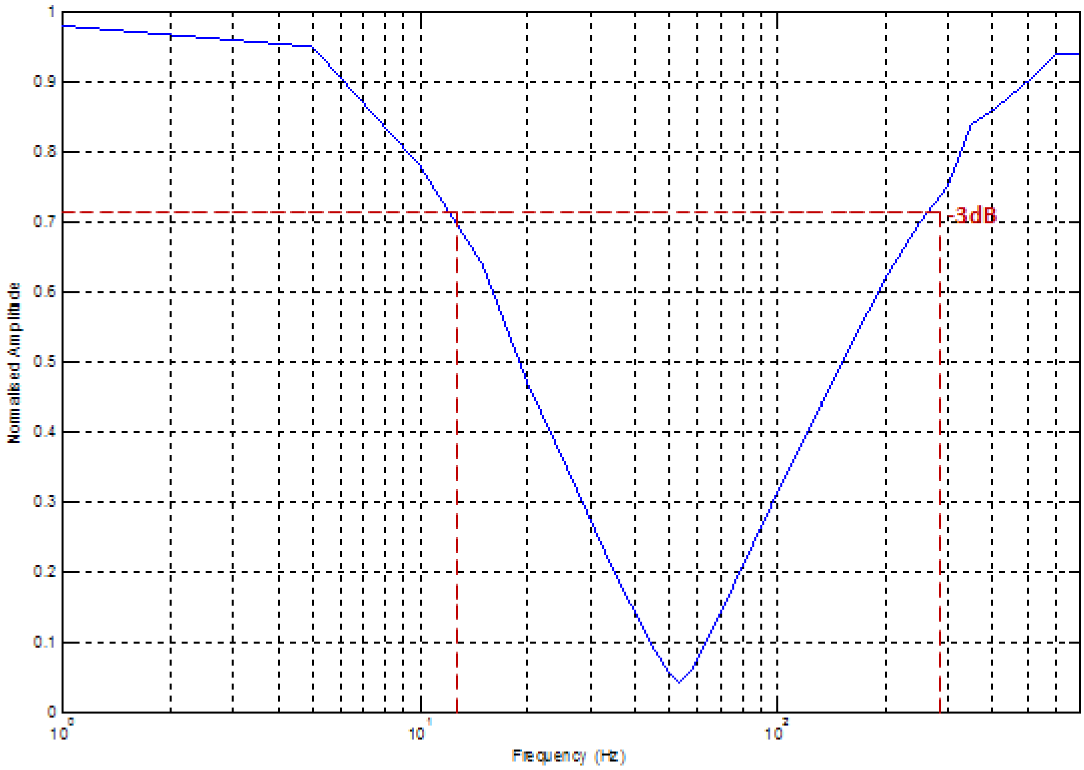


Figure 5-19 Frequency response of the notch filter

When this notch filter is applied to the ECG signals which have been processed by the high-pass filter and the amplifier, it can be seen from the oscilloscope that the profile of the ECG signals appears and the 50 Hz noise is largely suppressed.

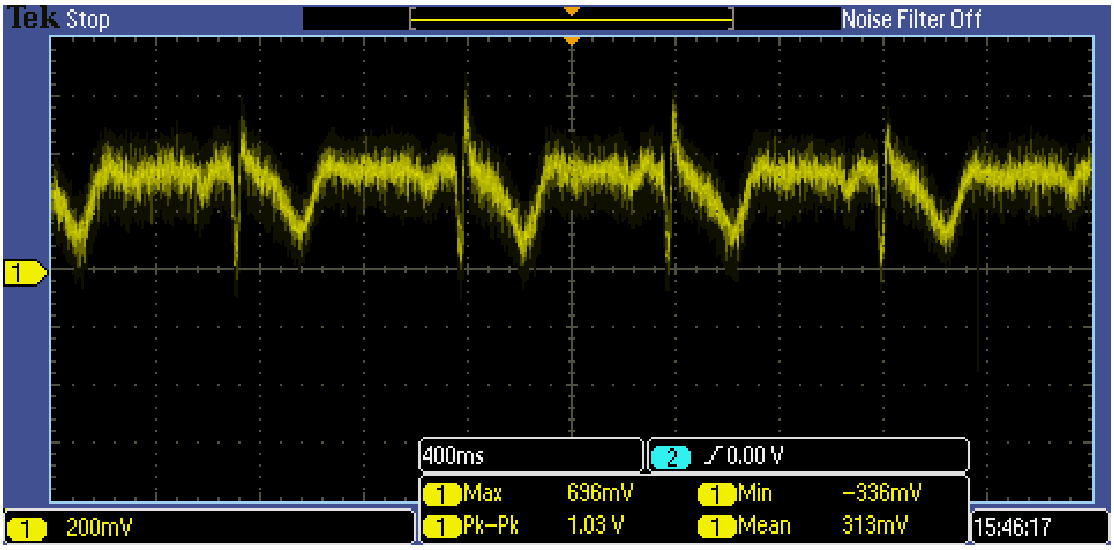


Figure 5-20 Result of the notch filter after the high-pass filter and the amplifier

The oscilloscope gives the maximum and minimum readings of the signals' amplitude (Max = 696 mV, Min = –336 mV), and the peak-to-peak value is 1.03 V. The frequency spectrum of the processed signals demonstrates the effect of the noise suppression, as shown in Figure 5-21. The normalised magnitude of 50 Hz harmonics is reduced from 1 to 0.053 by the notch filter.

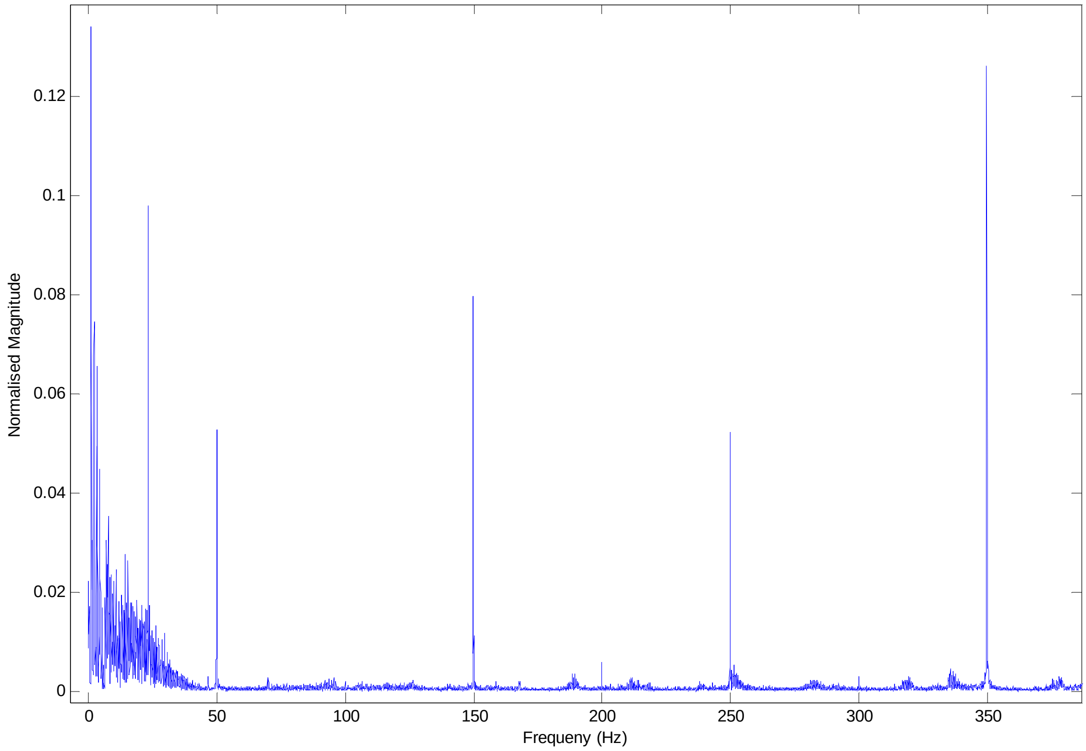


Figure 5-21   Frequency spectrum of the ECG signal after processed by notch filter

Dig into the data in Figure 5-20: the amplitude of the signal is from -336 mV to 696 mV. If further manipulation to the signals by the low-pass filter which is composed by single supply amplifiers, all parts of the ECG waveforms below 0 V will fail to reproduce. Hence, the voltage level shifting sub-module is introduced to preserve the integrity of the signals.

5.3.5 Voltage level shifting sub-module

The voltage level shifting circuitry performs two functions. First, since further filtering manipulation is undertaken using single supply amplifiers, all levels of the ECG signals should be shifted beyond zero volts without signal distortion. Second, the output of the signal conditioning module needs to fall into the range of the ADC's input voltage (i.e. 0 V to 3.3 V in this work). The signals from the output of the notch filter are fed into the input of the level shifting circuit, and the output was displayed on the oscilloscope. The result is given below in Figure 5-22.

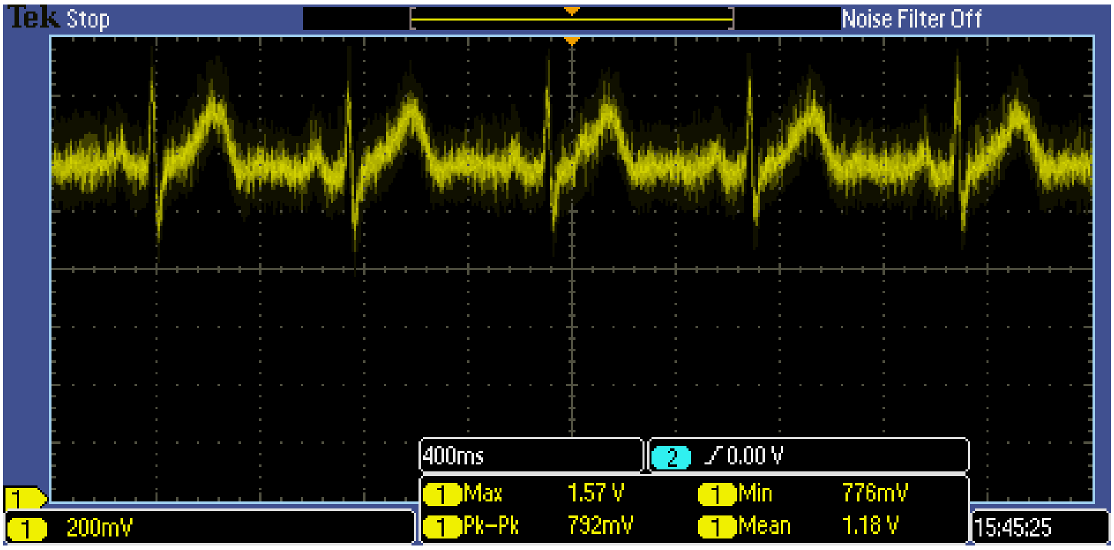


Figure 5-22 Result of the voltage level shifting circuitry

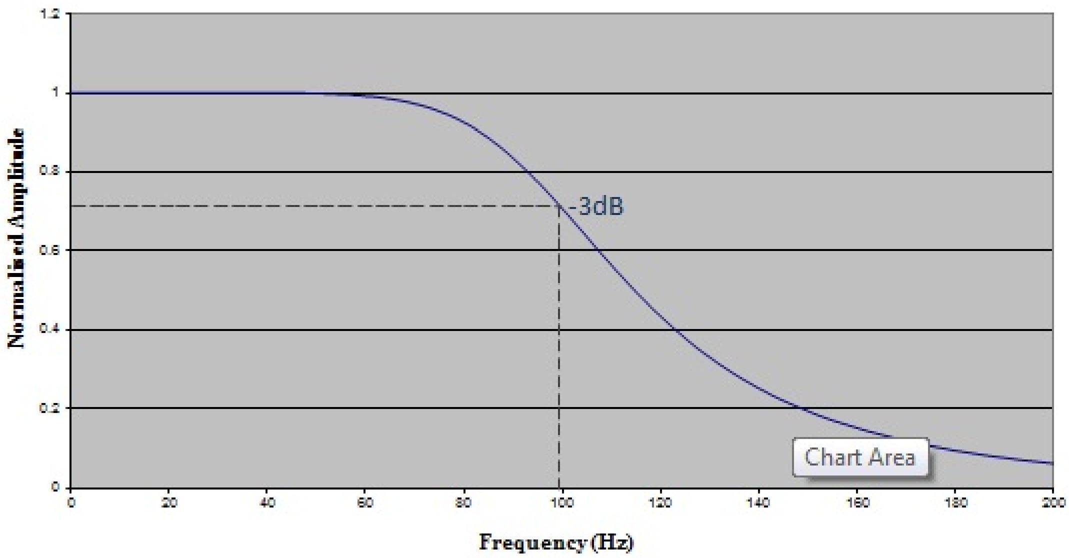
From observation, the baseline of the ECG signals is relocated and the waveform of the signal retains the original shape. The measurement demonstrates that the mean amplitude of the signals is 1.18 V. If this value is compared with the mean amplitude (313 mV) of the signal produced by the notch filter, it demonstrates that the level shifting sub-module moves the signals to a range of the ADC. The maximum level of the signal is 1.57 V, which are less than 3.3 V. The minimum is 776 mV, which exceeds zero volts. Thus, the signals are regulated to a proper amplitude range and level.

5.3.6 Result and performance of the low pass filter

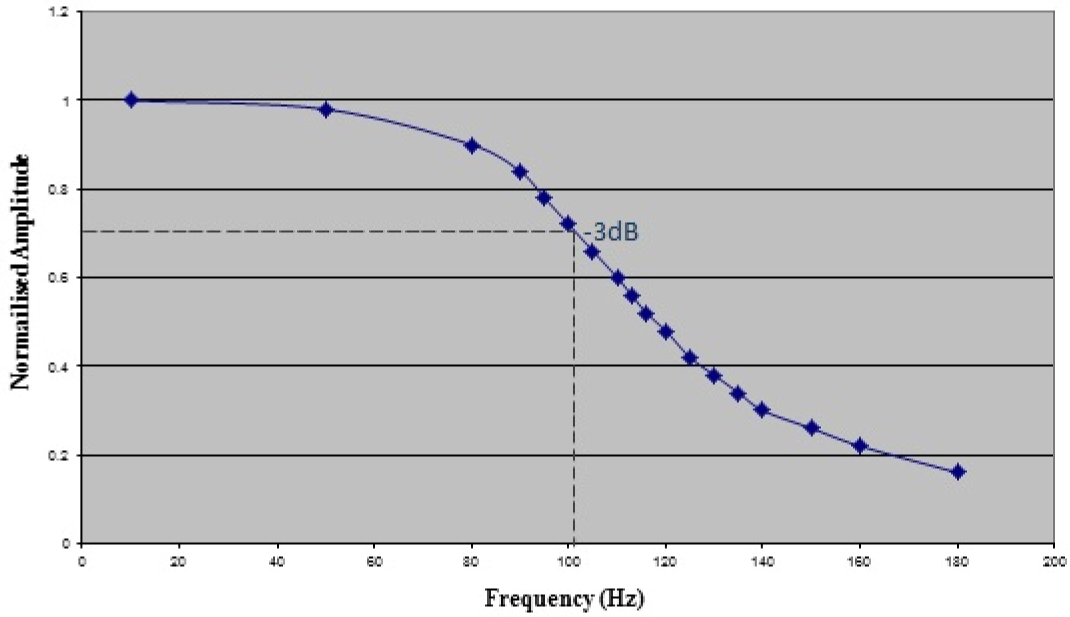
Although the 50 Hz interference was suppressed, it can be observed from Figure 5-21 that the harmonics of the noise components beyond 100 Hz still exist and contaminate the ECG signals.

To counter the noise disturbance, the 4th-order low-pass Butterworth filter with 100 Hz cut-off frequency is used to attenuate the high frequency noise without the loss of signal integrity from a clinical perspective. This filter is constructed by two-stage Sallen-Key architecture with unity-gain. An experiment was performed to characterise its frequency response, using a 2 V peak-to-peak swept frequency sine wave. The output of the filter was fed to an oscilloscope and the amplitudes were recorded as a function of frequency.

The normalised amplitude results were plotted as a function of frequency. Moreover, the run-time frequency response was compared against the design performance, using the software ‘Filter_Pro’ by Texas Instruments. The plots are shown in Figure 5-23 (a) and (b).



(a)



(b)

Figure 5-23 (a) Designed frequency response of the low-pass filter

(b) Run-time frequency response of the low-pass filter

While this low-pass filter is applied to the output of the voltage level shifting circuit, the outcome was obtained using the oscilloscope. The screenshot is presented in Figure 5-24.

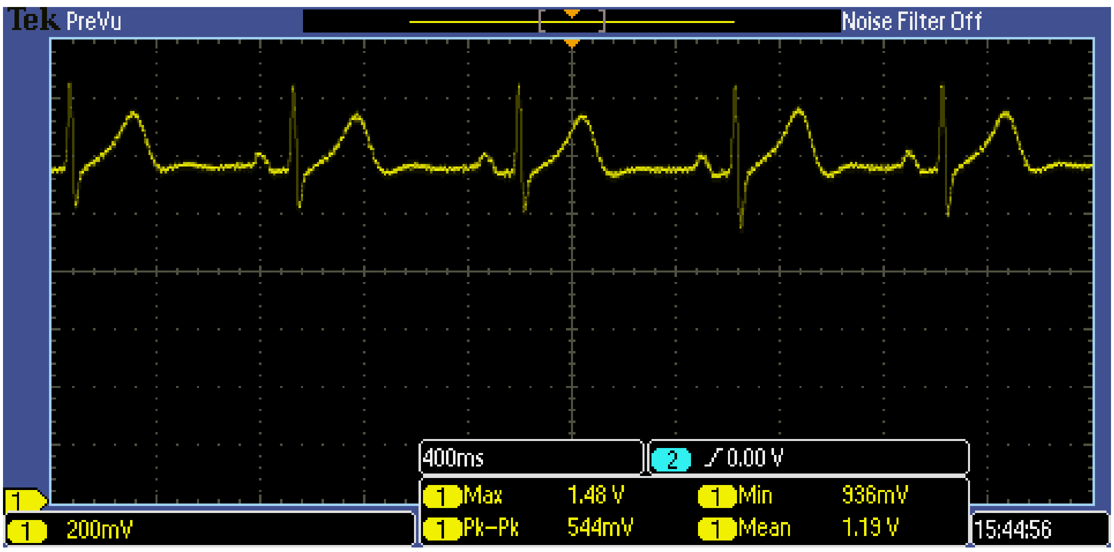


Figure 5-24 Result of the low-pass filter

The oscilloscope measurement confirms the peak-to-peak amplitude of the filtered signal is 544 mV. The signal's voltage range can be read as maximum 1.48 V and minimum 1.19 V, which is still in compliance with the ADC's input scope. The frequency spectrum of the processed ECG signal clearly confirms that almost all noise harmonics are suppressed by the multiple filters. The FFT analysis of the processed ECG signal is shown in Figure 5-25. The FFT analysis of the restrained noise in the signal is illustrated in Figure 5-26.

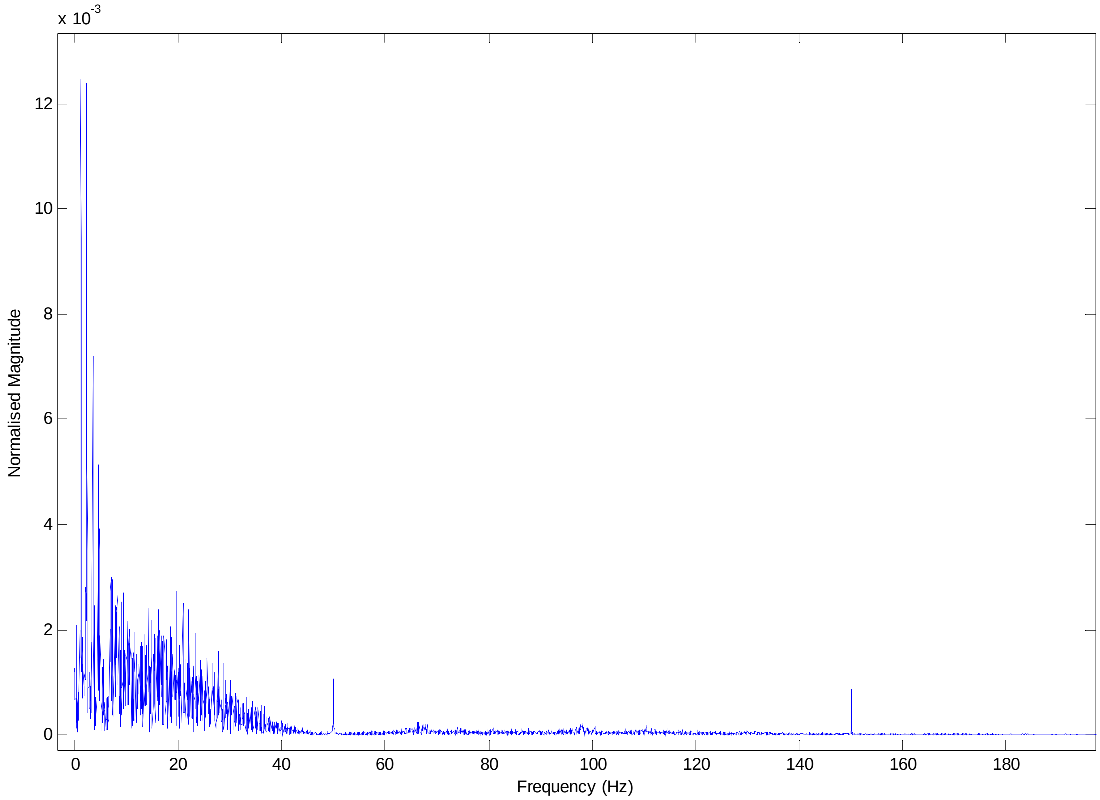


Figure 5-25 Frequency spectrum of the ECG signal after being processed by multiple filters

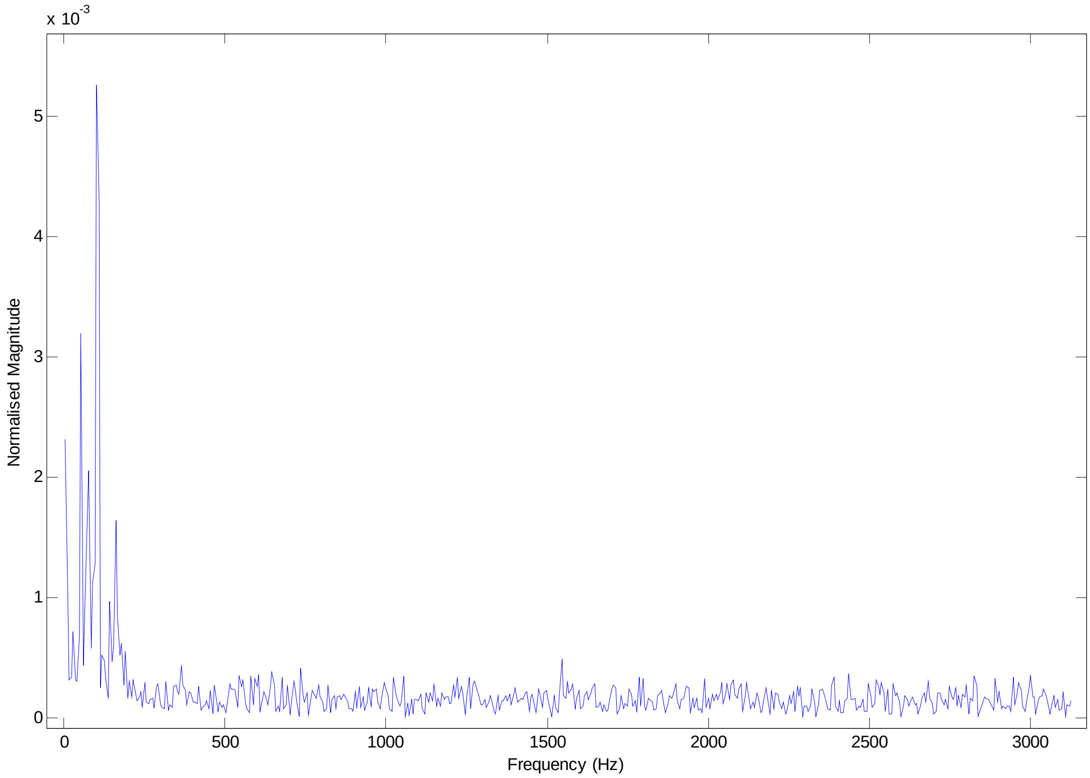
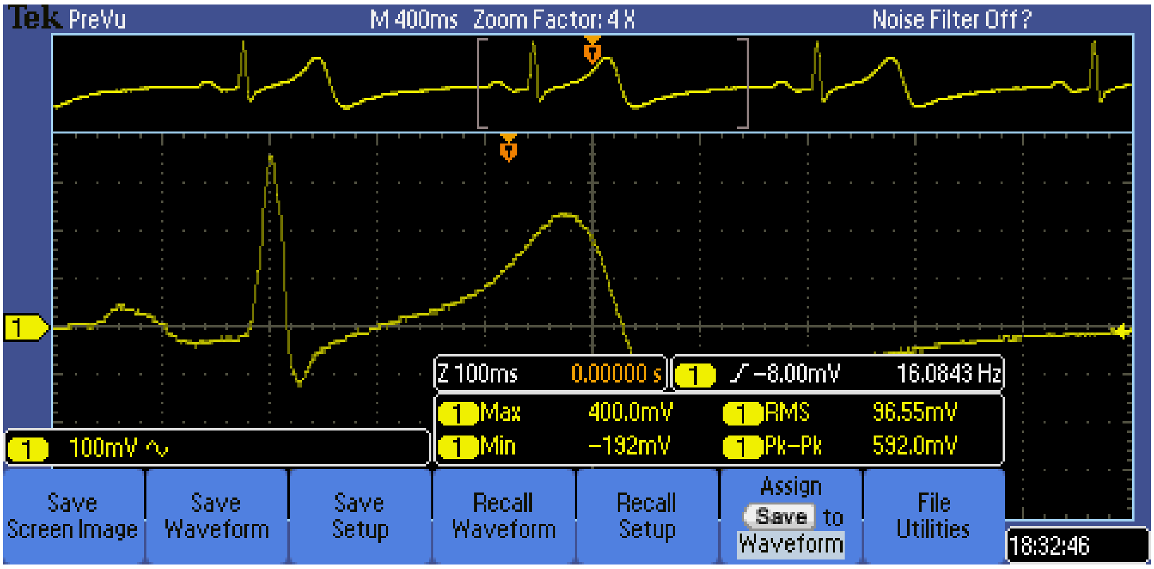


Figure 5-26 Frequency spectrum of the suppressed noise

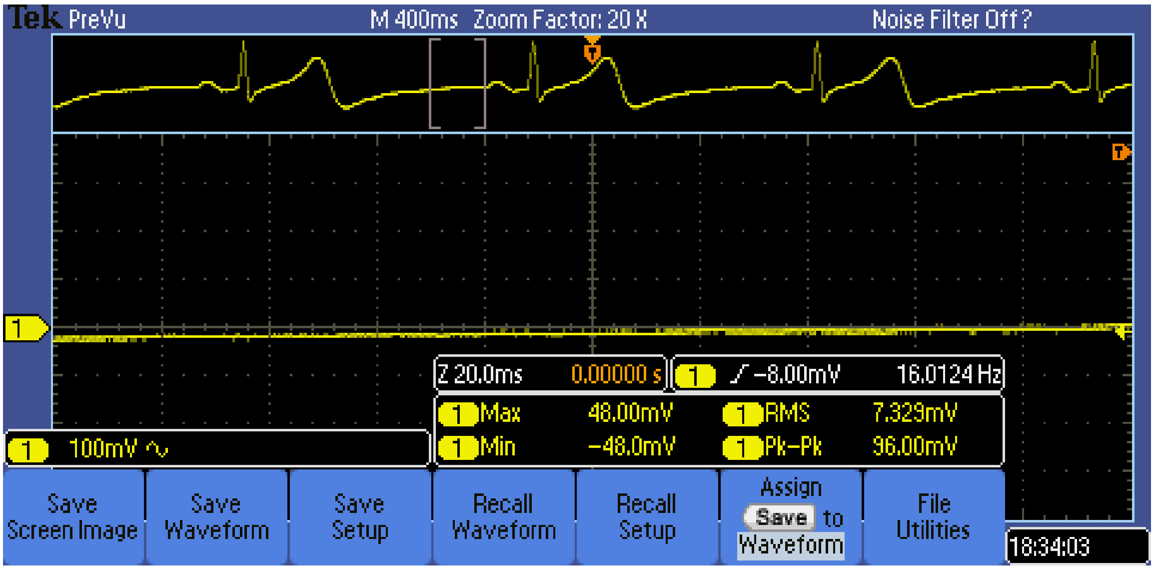
This design employs the filters of no more than 4^{th}-order. Indeed, higher order analogue filters can yield better frequency response performance to attenuate the noise components which exceed the pass band. However, high-order analogue filters can lead to phase distortion of the signals, which should be avoided in biomedical applications (Gaydecki, 2004). Therefore, a trade-off ought to be considered between complexity of filter design (such as cascading) and performance. From the results and the performance analysis, these filters are able to meet the requirements for the recovery of the ECG signals.

A signal-to-noise ratio analysis reveals how the quality of the ECG signals is improved after multiple filtering. The SNR calculation can be estimated by calculating the root mean squared (RMS) values of the ECG signal, and obtaining the ratio.

ECG signals were acquired and gated measurement is applied using a Tektronix DPO 2012 digital oscilloscope to obtain the RMS values of the ECG signal and the noise. AC measurement was employed to eliminate the effect of DC offset to the RMS. The ECG signal applied to the oscilloscope had been pre-filtered leaving a band between 0.5 Hz and 100 Hz, with a notch at 50 Hz, as discussed previously. The timebase was set to show a full ECG complex, to enable an accurate measurement of its RMS value. The RMS value of the signal was performed by the oscilloscope and obtained from the screen as shown in Figure 5-27 (a). In this example shown, the ECG signal had a peak-to-peak amplitude of 592.0 mV and its RMS was 96.55 mV.



(a)



(b)

Figure 5-27 RMS values measurement in vivo. (a) Result of ECG signal RMS measurement, (b) Result of concomitant noise measurement

When measuring the RMS of the noise, the ECG interval period is considered as concomitant noise without the signals. To assess it accurately, and to obtain a longer quiescent period, the tested subject should be static and calm. The result of one such test is shown in Figure 5-27 (b). The noise was 7.329 mV RMS in amplitude with 96.00 mV peak-to-peak amplitude. Thus, the SNR of the processed ECG signal in this case is:

$$SNR = 20\lg\frac{RMS_{signal}}{RMS_{noise}} = 22.39dB \qquad (5\text{-}6)$$

Where RMS_{signal} is 96.55 mV, RMS_{noise} is 7.329 mV.

In reality, the measurement of the RMS_{signal} value is not strictly true as the reading of the signal also includes some noise. However, the concomitant noise has been suppressed and the deviation of the value is not particularly large. In view of this point, the tolerance of the slight difference is acceptable and the ratio still applies to the feature of the signal.

At this stage, the morphological representation of the ECG signals was achieved by the front-end analogue signal conditioning module. The processed ECG signals are satisfactory for clinical monitoring purpose and are ready to be digitised and translated towards the networks.

The sampling rate of the ADC is set at 200 Hz. There is both a merit and disadvantage between, on the one hand, oversampling, which can preserve the higher frequency components of the signals, and on the other, transmission capacity depending on the network stream flow. A series of studies have indicated that a sampling rate of four to six times the upper limit cut-off frequency of the signals is recommended for adults (Kligfield. et al., 2007, Goldberger. and Ng., 2010). However, under specific transmission conditions including the Internet flow-rate, the data which are generated from sampling conversion should ensure delivery to the clients without data loss or network congestion.

5.3.7 Dual-channel implementation

This system is designed to perform multi-channel monitoring of the user. In this experiment, two channels have been developed (Lead I and Lead II ECG signals) so as to prove the feasibility of the multi-channel capability of the system. By similar means, extended channels may be applied to the system. The Lead I and Lead II ECG signals are using the PCB board and are displayed on the oscilloscope, as shown in Figure 5-28.

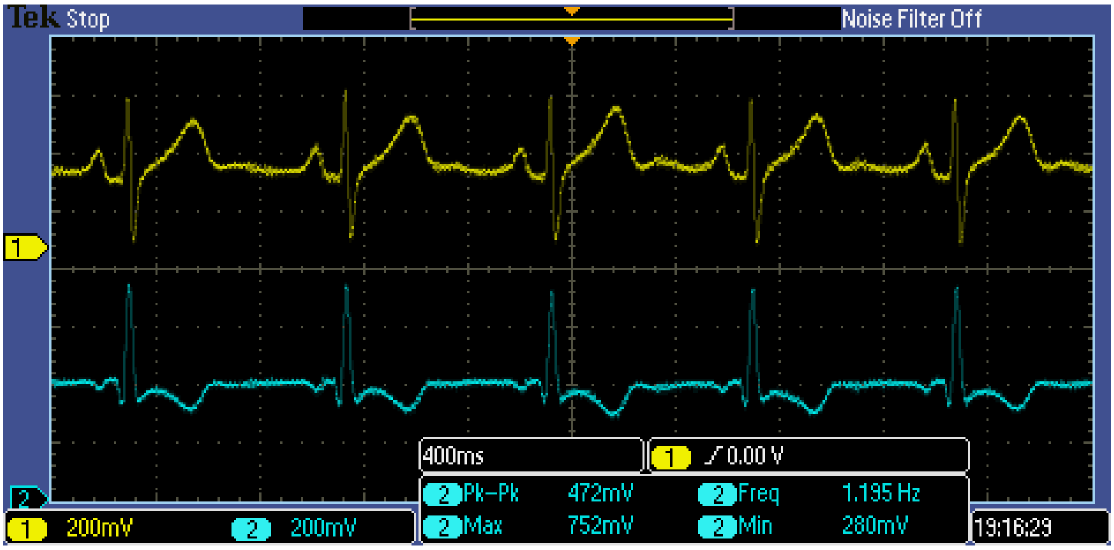


Figure 5-28 Dual-channel of the Lead I and Lead II ECG signals on the oscilloscope

In the figure, Channel 1 (yellow) represents Lead II ECG signals and Channel 2 (cyan) displays Lead I ECG signals. On the strength of these results, it is definite that the usage of the front-end PCB board accomplishes the functions of the dual-channel and the signal conditioning module.

5.3.8 Respiration

The respiration signals are detected *in vivo* and the experimental subject is always erect without noticeable body motion through the experiment. In order to have correct performance, the belt respiration transducer should be worn around the chest and adjusted using the inelastic belt attached to the sensor strap. Proper tension adjustment of the strap is an important factor to ensure accuracy. When the transducer is initially fitted, a deep breath is necessary to ensure that the maximum extension of the sensor strap can be accommodated. A typical output signal of the sensor, taken from an oscilloscope, is shown in Figure 5-29.

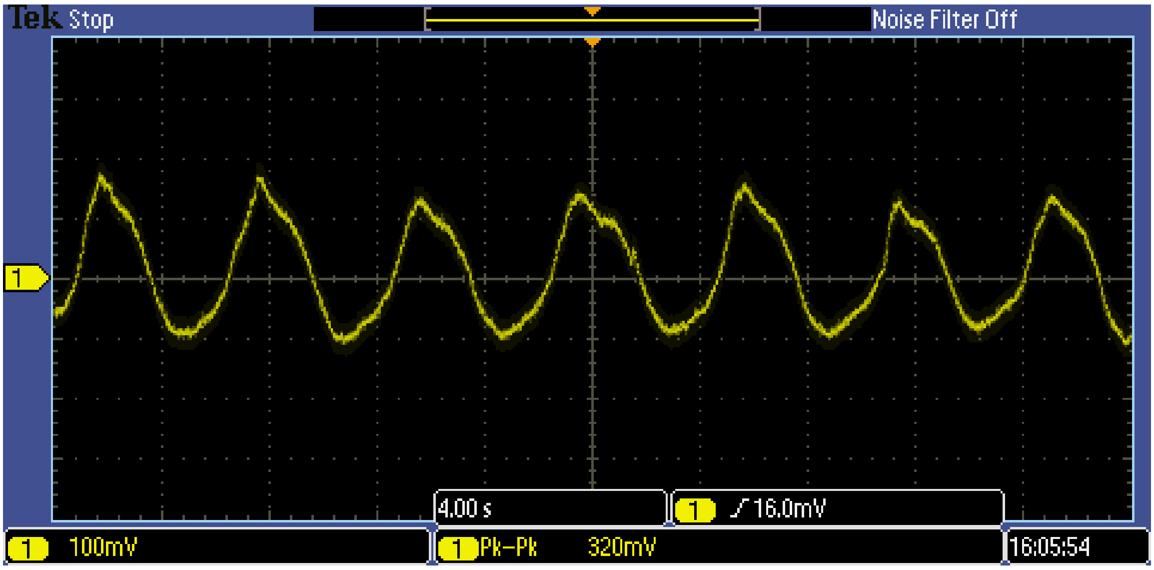


Figure 5-29 The initial analogue respiration signals displayed in the oscilloscope

By measuring changes in thoracic circumference using this transducer, the breathing patterns or respiration signals can be clearly characterised and are ready to be dispatched by data transmission module through the Internet. The respiration signals will be transmitted in real-time together with other biomedical signals in the multiple-sensor mode.

5.3.9 Body temperature

The body temperature measurements were obtained *in vivo*. “Normal” body temperature values are generally given for an otherwise healthy, non-fasting adult, dressed comfortably, indoors, in a room, during the morning but not shortly after arising from sleep. The readings were obtained in a normal room with a temperature ranging between 22.7 and 24.4 $^{\circ}$C. The temperature sensor was attached in the axilla of the subject. It should be noted that there is sensor respond time during the measurement and the time constant of the device is approximately 4 seconds. However, body temperature fluctuations have an affect over several minutes. Therefore, the values were read first from a multimeter until a stable reading was observed. These voltage readings were converted to temperature values according to Equation 4-2. A series of body temperature from the subject were recorded through the daytime (9:00 to 21:00), shown in APPENDIX D and plotted in Figure 5-30.

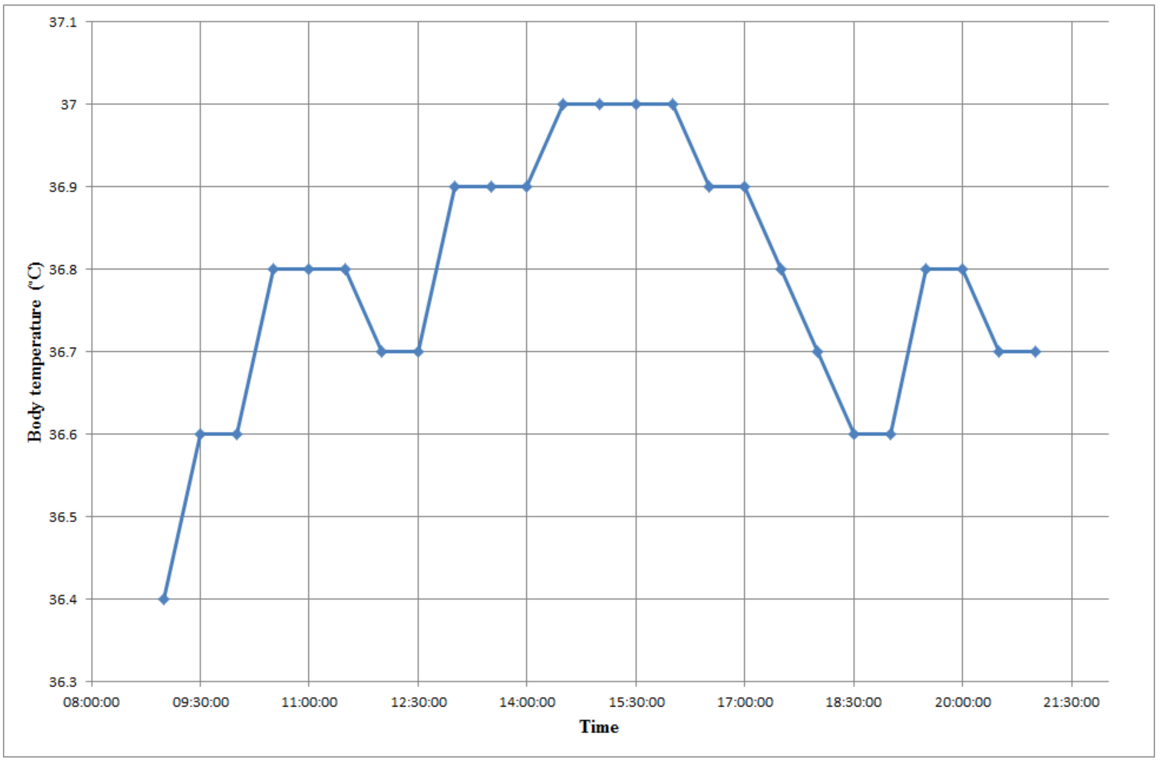


Figure 5-30   Body temperature records through the daytime

Figure 5-30 intuitionally depicts that the body temperature fluctuate over the daytime is approximately within 0.6 °C. A comparison between the sensor results and the reference values read by the thermometer shows that an almost constant bias is present between the sensor measurement and the thermometer measurement. This bias may be introduced by several reasons, such as sensor header manufacturing process, sensor leads, reading moment, which can be calibrated according to Equation 5-1. The bias of the sensor measurement was fairly removed prior to display. The body temperature values were to be transmitted together with the ECG and respiration signals in synchrony over the Internet.

5.4 Results: transmission via the Internet

Once the multiple biomedical signals are acquired and processed by the front-end PCB board, they are ready to be delivered through the networks. The multi-channel data are transferred via either the Ethernet or the Wi-Fi onto the Internet. The transmitted signals can be received and displayed by using the dedicated LabVIEW program at the client end anywhere with the Internet connectivity. This section will be discussed how the ECG signals, heart rate, respiration signals and body temperature are displayed synchronously, in near real-time (delay is less than one second), on the LabVIEW program at the client end. An analysis of ECG signal behaviour variation under different physiological conditions will be given based on the experiment *in vivo*. Furthermore, the impact on the signals in relation to different network speed will also be explored in the experiment.

5.4.1 The multi-sensor signals represented in the LabVIEW software

The multiple biomedical signals were acquired, processed and transmitted via the Internet by the system. These are ultimately graphed by the display module at the client side. The signal display function is fulfilled by means of the LabVIEW program.

Prior to operation, the display module must be configured correctly to receive the signals from the object via the Internet. An IP address must be provided (in the dialog box 'Http URL') in order to orient the object. When the start button  is clicked, the live signals are plotted; the function is ended by clicking the stop button 

. Figure 5-31 shows a sample real-time ECG signals from the LabVIEW program.

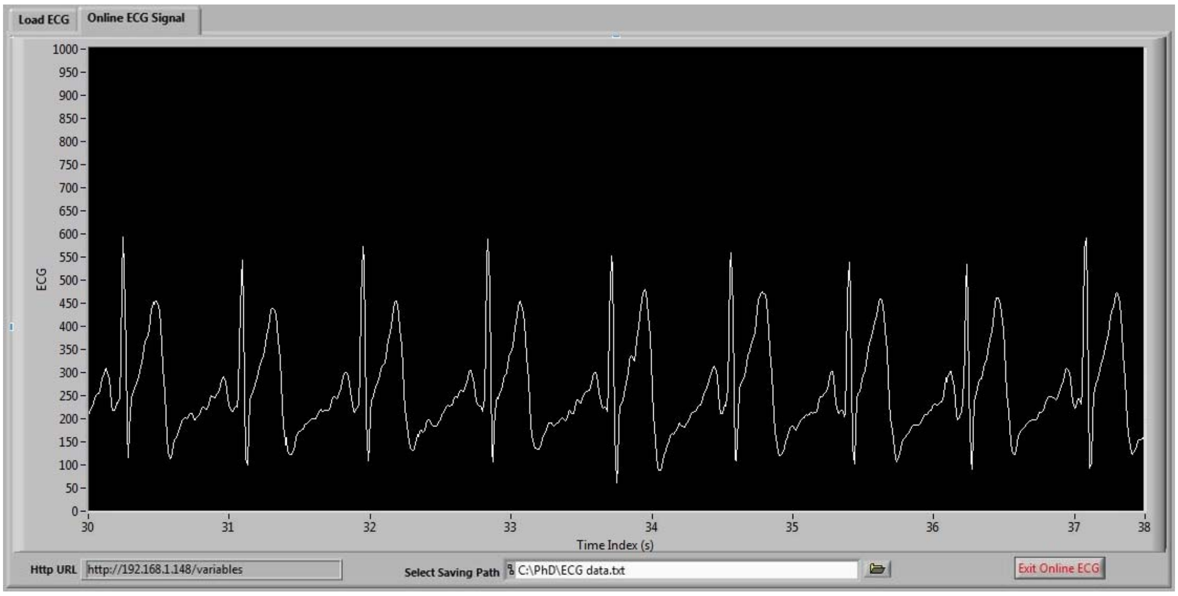


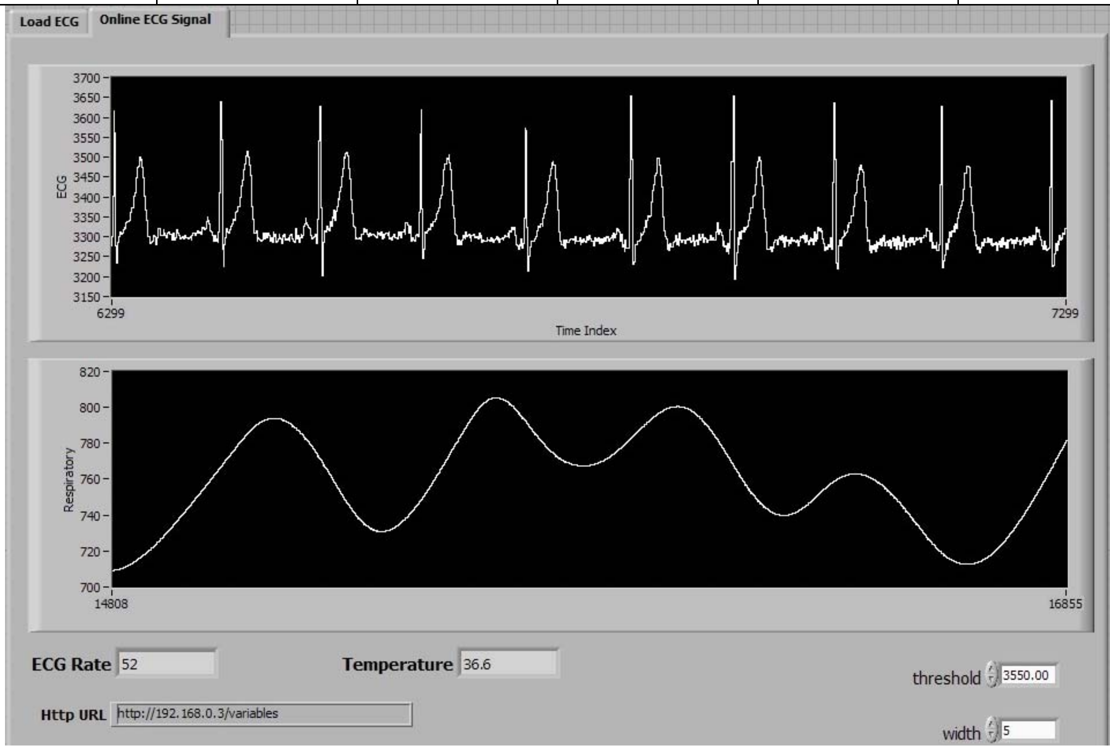
Figure 5-31 Screen view of an ECG signal produced by the LabVIEW program

The ECG signals are displayed in the window 'Online ECG Signal', the amplitude range of which lies between 100 mV to 600 mV approximately. By comparison, the ECG signal before transmission, shown in Figure 5-24 is noisier. This is because the LabVIEW program performs real-time filtering of the signal using a Savitzky-Golay digital filter, prior to display. Onto this stage, the vital signals can be displayed and monitored in real-time by employing this program.

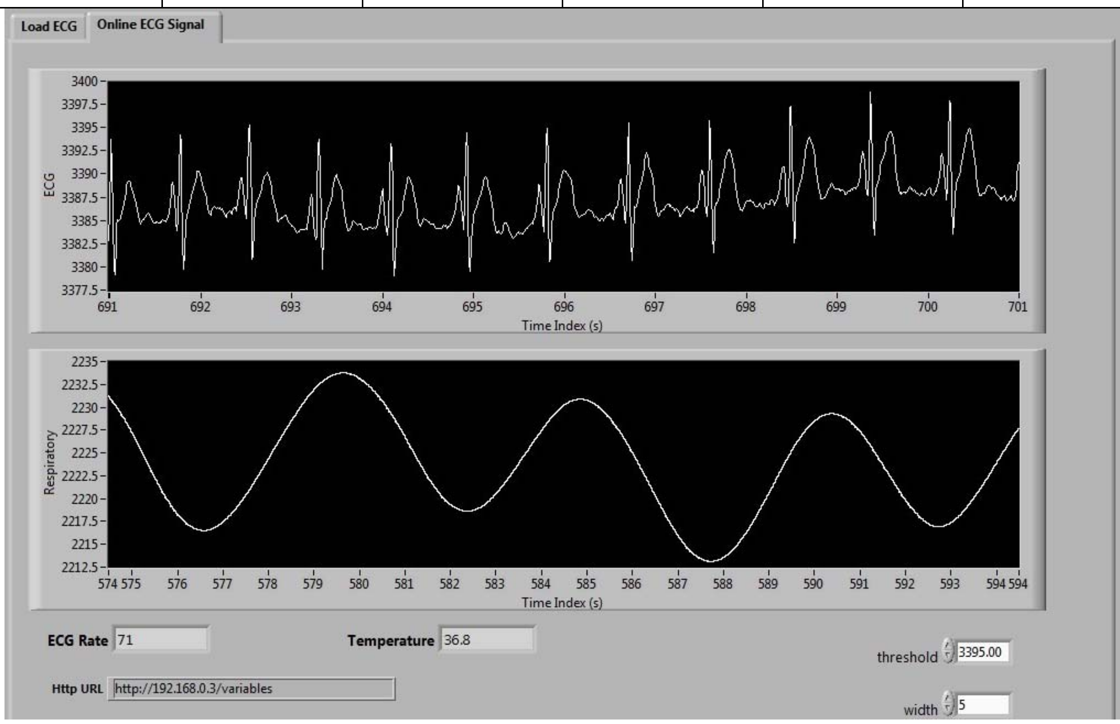
In the experiment, the various physiological signals were acquired from 8 volunteers *in vivo*; University ethics committee approval had been sought and given for this activity (see APPENDIX F). Eight elective volunteers were adults (two female, six male) ranged in age from 23 to 41 with an average age of 31.6 years and an average weight of 70 kg. All volunteers were non-smoker having no cardiac disease history. All experiments were taken in the daytime from 9am to 5pm. The volunteers wore the ECG electrodes, the Pneumotrace II model 1132 respiration transducer and the body temperature sensor. Real-time ECG signal, heart rate, respiration signal and body/skin temperature were measured no less than one minute. During the test and measurement, the subjects stand in an upright position and remain in a static state. The system operates in wireless mode and is connecting the Internet. The following table lists some experimental results obtained from the volunteers, together with their personal physiological information.

Table 5-5 The results of the real-time multi-sensor signals displayed in the LabVIEW program GUI

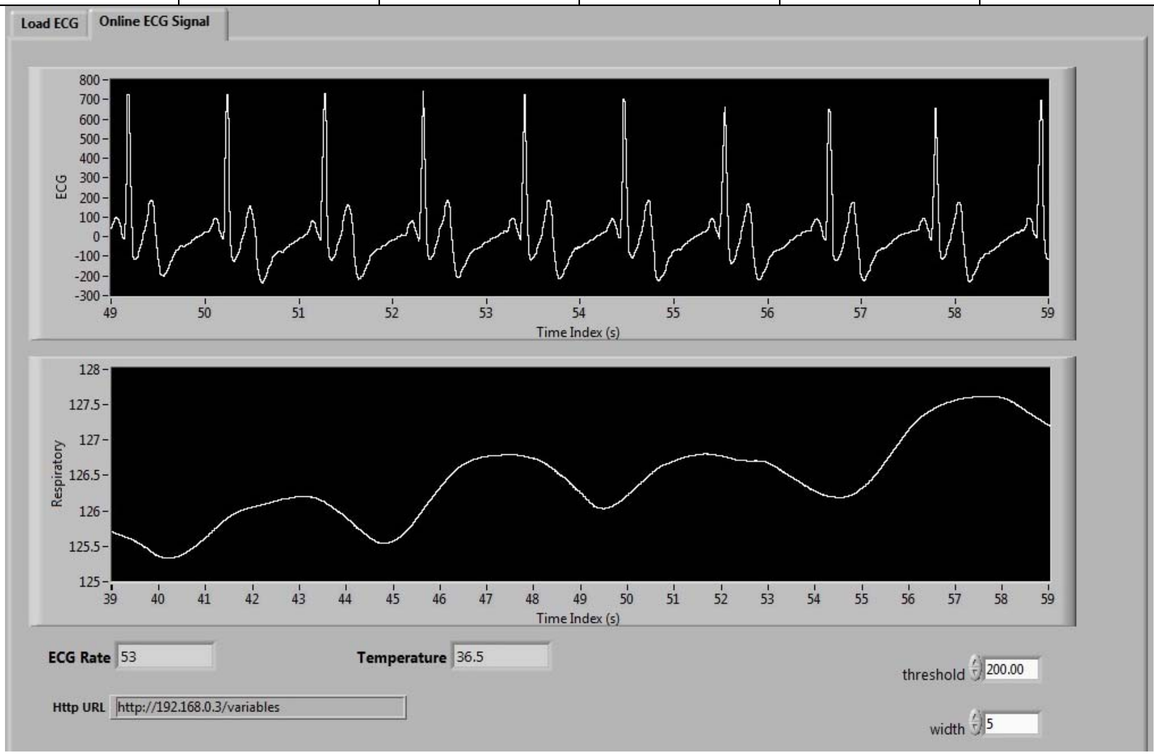
Volunteer 1	Gender: male	Age: 32	Weight: 71 Kg	Heart rate: 52 BPM	Temperature: 36.6 ˚C



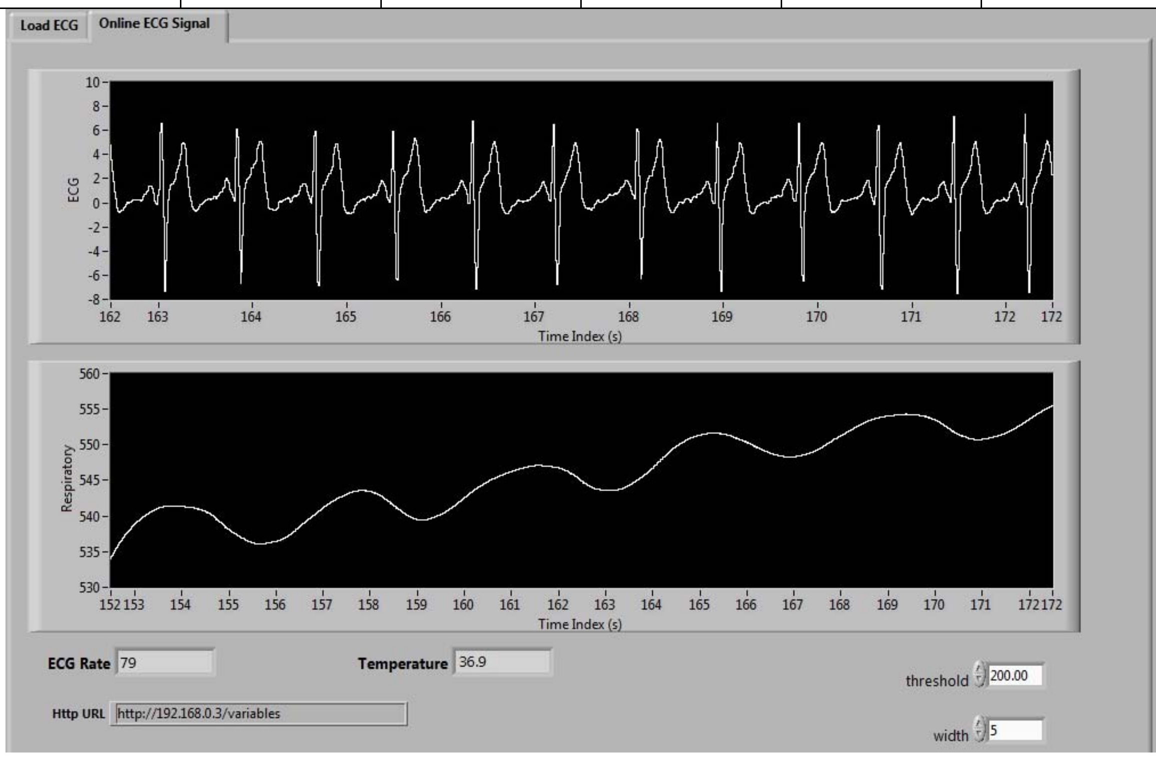
Volunteer 2	Gender: male	Age: 24	Weight: 82 Kg	Heart rate: 71 BPM	Temperature: 36.8 ˚C



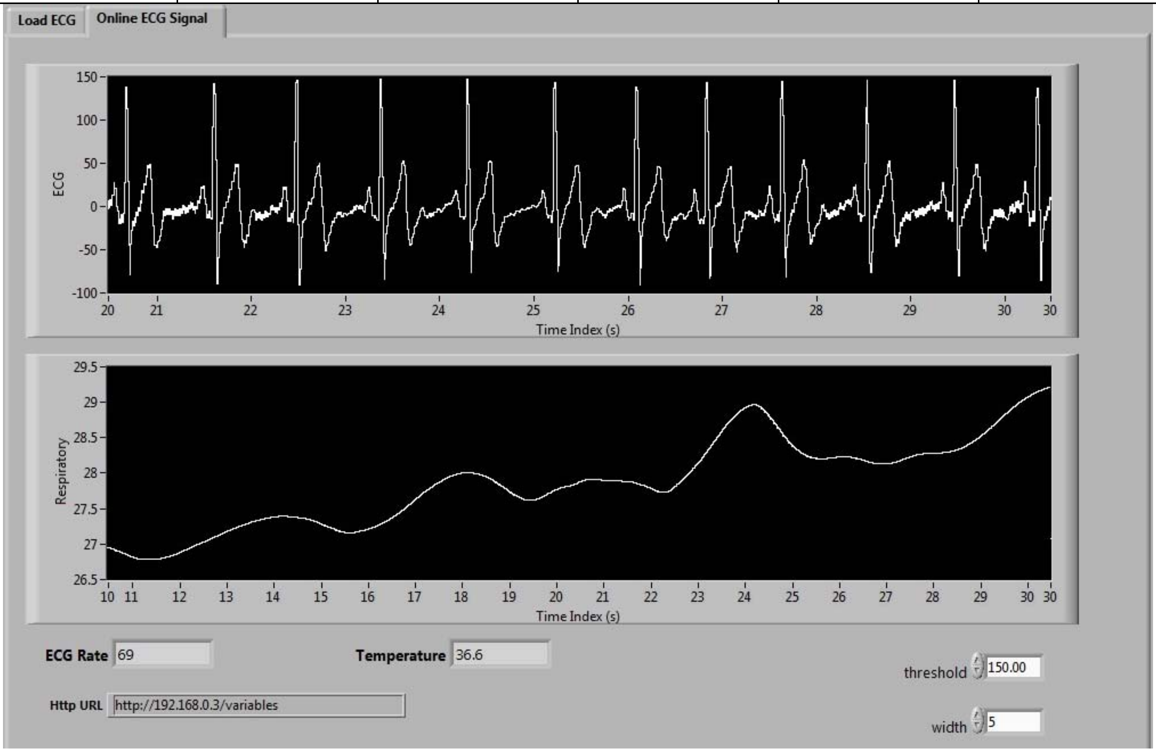
Volunteer 3	Gender: male	Age: 39	Weight: 80 Kg	Heart rate: 53 BPM	Temperature: 36.5 ˚C



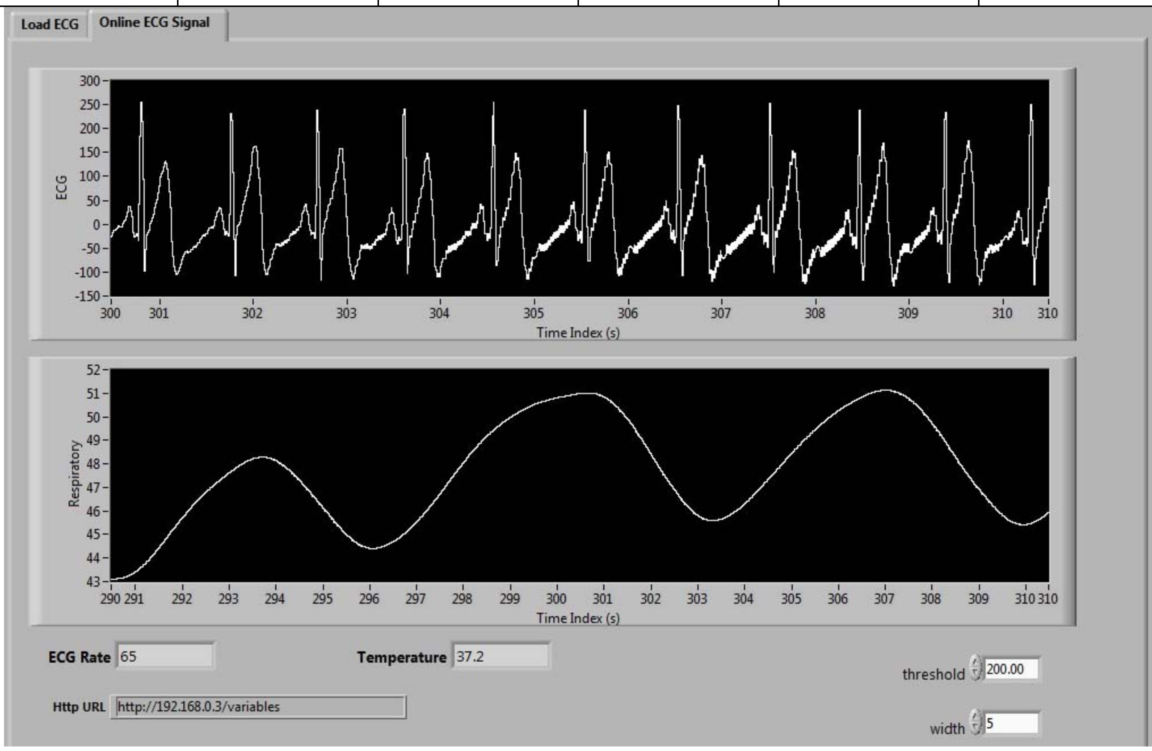
Volunteer 4	Gender: male	Age: 41	Weight: 75 Kg	Heart rate: 79 BPM	Temperature: 36.9 ˚C



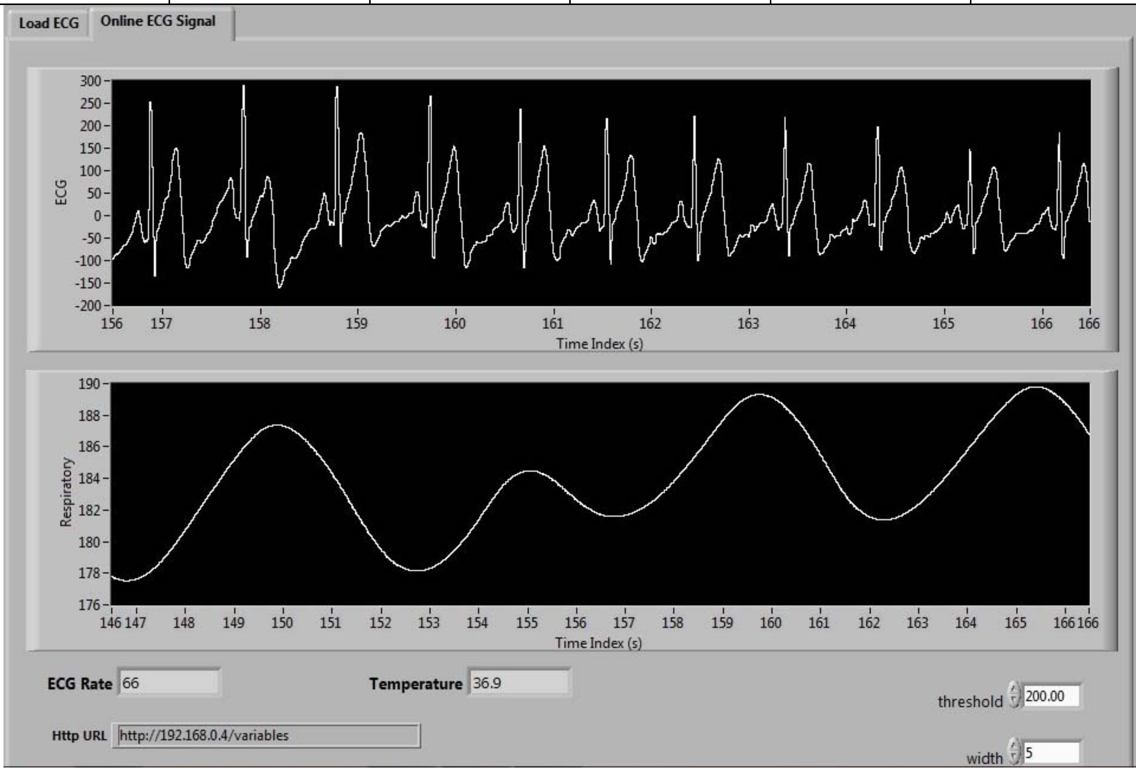
Volunteer 5	Gender: female	Age: 23	Weight: 58 Kg	Heart rate: 69 BPM	Temperature: 36.6 ˚C



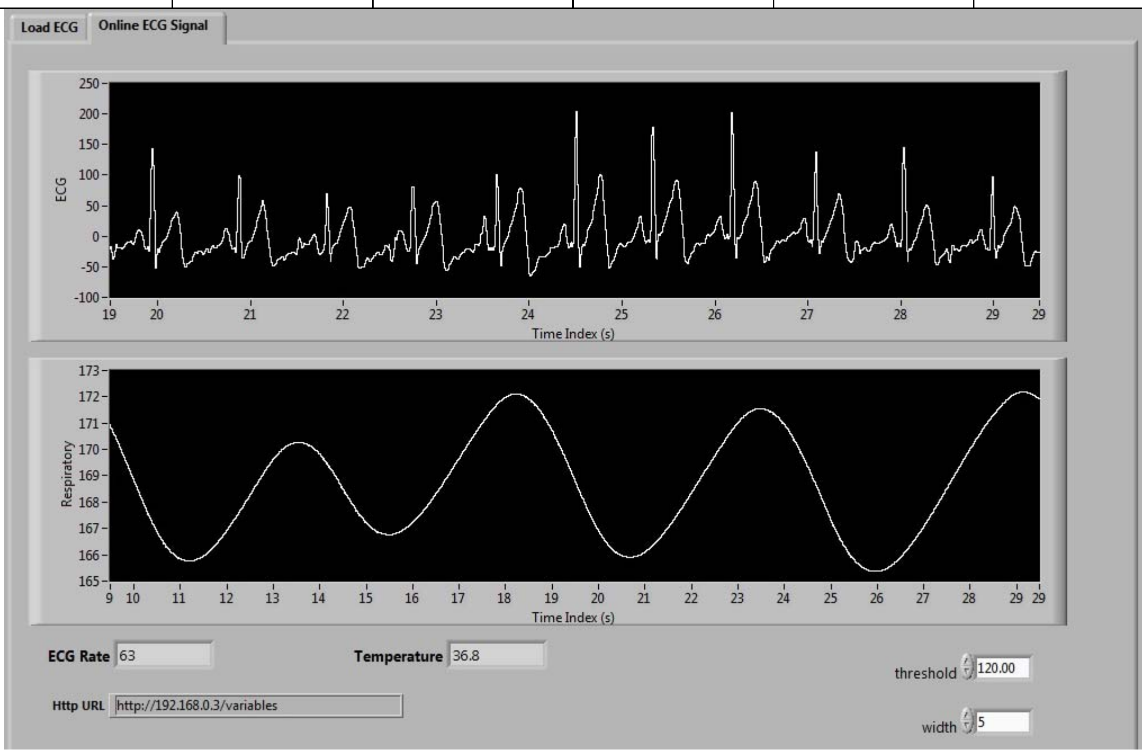
Volunteer 6	Gender: male	Age: 36	Weight: 77 Kg	Heart rate: 65 BPM	Temperature: 37.2 ˚C



Volunteer 7	Gender: male	Age: 25	Weight: 62 Kg	Heart rate: 66 BPM	Temperature: 36.9 ˚C



Volunteer 8	Gender: female	Age: 33	Weight: 55 Kg	Heart rate: 63 BPM	Temperature: 36.8 ˚C



As confirmed by the results in the table, the real-time ECG, respiration and body temperature signals are acquired and displayed by the LabVIEW program in synchrony. The heart rate based on the ECG signals is calculated simultaneously. A clock stop was used to measure the delay time of signal transmission over the Internet. When an event occurred during signal acquisition, e.g., an amplitude adjustment, the time elapsed is recorded whilst the event appeared in display. The transmission delay of the data from acquisition to the clinician display was less than one second, almost unnoticeable.

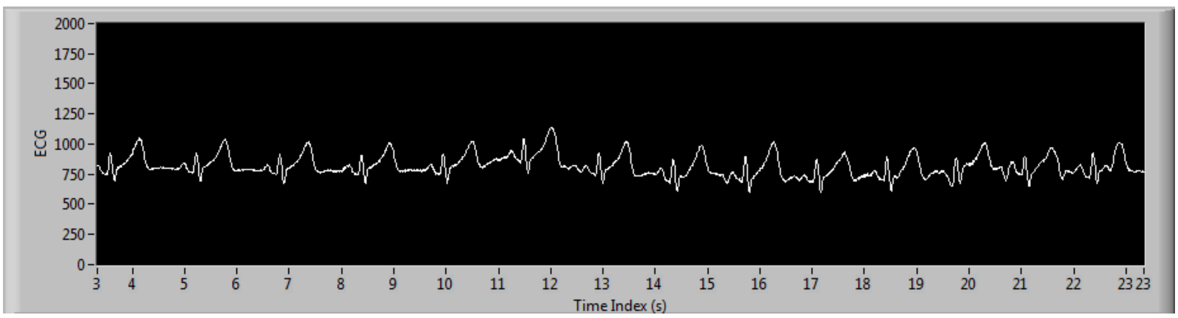
5.4.2 Analysis of system performance under various physiological conditions

It is well known that a subject's motion may introduce artefacts in relation to biomedical signal capture. As illuminated in the previous chapter, the effects of motion artefacts are most severe with respect to electrophysiology signals, such as ECG signals.

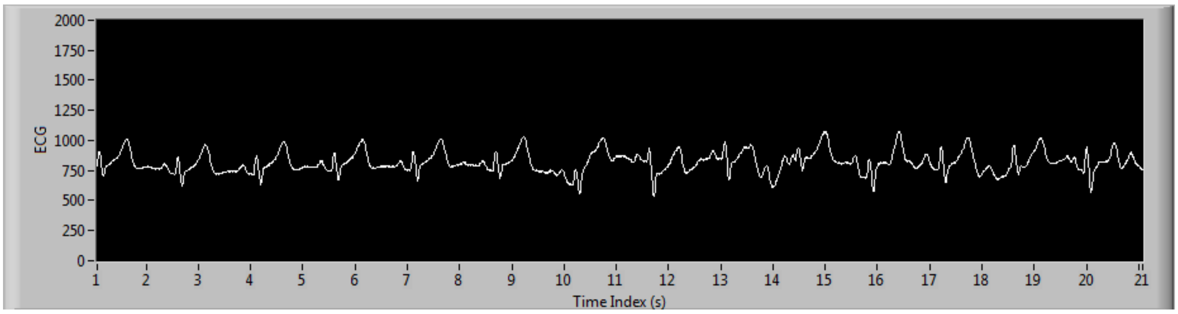
An experiment was undertaken to determine the robustness of the system performance in relation to movement artefacts. This included walking, arm movement, deep breathing, shivering and coughing. Brief overview of the results is given in Table 5-6.

Table 5-6 The ECG signal obtained under the different physiological conditions

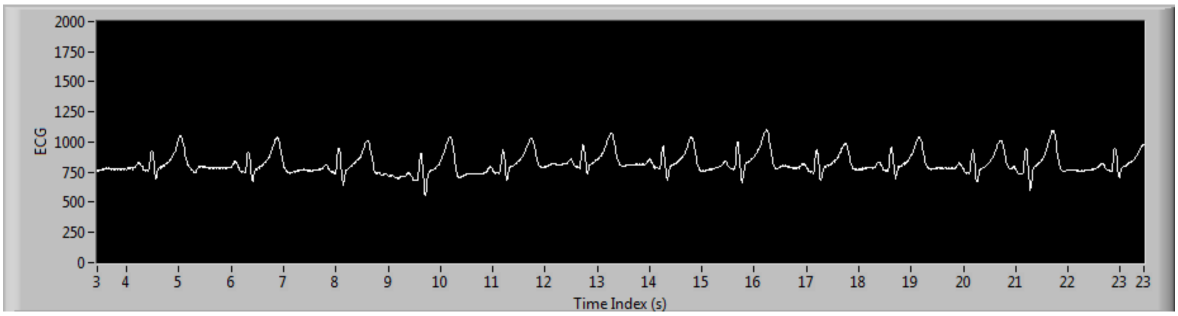
Walking



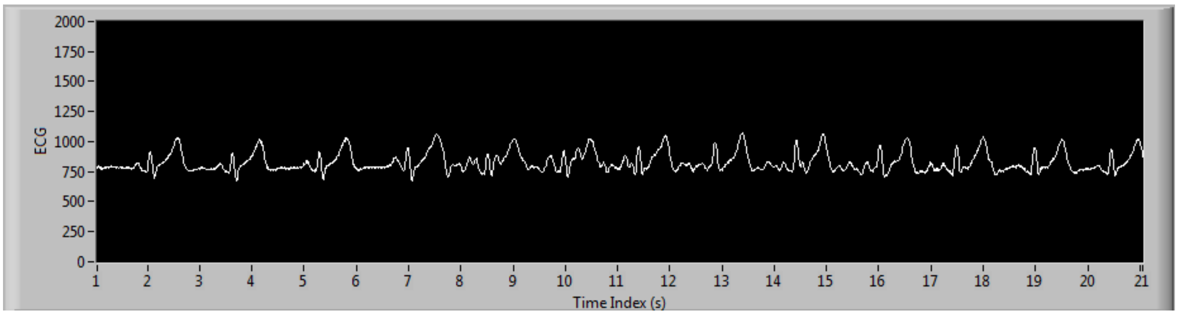
Arm movement



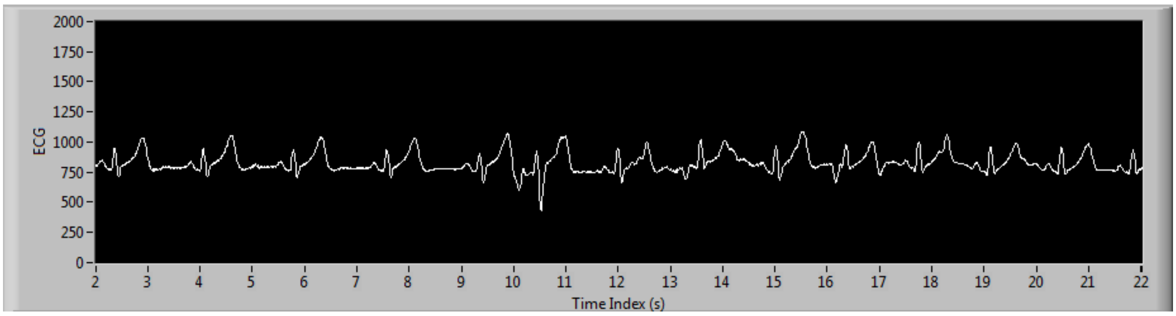
Deep breathing



Shivering



Coughing



From the results in above table, the real-time ECG signals manifest mild base line drift whilst walking during the signal acquisition; however, the quality of the ECG signal is still acceptable. If the measurement is taken while vigorously moving the arms, the signals appear considerably distorted. In contrast, deep breathing has little effect on the system's performance. In respect of shivering, the ECG signals were contaminated with spike-like interference but with less baseline wandering. Coughing induces transitory interference, i.e., when the cough occurs.

To generalise these experimental results, motion artefacts mainly give rise to baseline drifting (relevant low frequency noise) and sharp spike-like interference (relevant high frequency noise). As the designed system facilitates this interference by employing both high-pass and low-pass filter to cope with them, the noise does not affect the ECG waveform drastically.

5.4.3 Analysis of system performance at different Internet speeds

The Internet proffers backbone network for the information distribution in this work. The network speed is an important factor governing the system performance in relation to data transmission. To explore this, simulated ECG signals from the Fluke MPS450 Multi-parameter Simulator were acquired and transmitted by the system at different Internet network speeds. The network speeds were pre-set as 100 Mbps, 54 Mbps and 36 Mbps in this experiment. Portions of these ECG signals, transmitted at these different speeds and received by the LabVIEW program, are shown in Table 5-7.

Table 5-7 The ECG signals in the LabVIEW program at different Internet speed

Internet speed: 100 Mbps

Internet speed: 54 Mbps

Internet speed: 36 Mbps

The ECG signals were recorded for 30 seconds under different network speeds in the experiments and segments of the stored data are here displayed. At 100 Mbps, the ECG signals are reproduced correctly. At 54 Mbps network speed, the signals manifest some data conflict once or twice within 30 seconds. With an Internet speed of 36 Mbps, the data collision occurs slightly more frequently but signal interpretation is still possible. From the ECG signal interpretation in the GUI, it is noticeable that the integrity and continuity of the received signal is strongly dependent on network speed. Moreover, the computer which runs the LabVIEW program should meet the system requirements for the LabVIEW development system, listed in Appendix E.

5.5 Storage and retrieval

Recording the real-time biomedical signal data in the client-side computer is essential to the clinicians for further handling. The storage function has been incorporated in the LabVIEW software. Furthermore, users can recall the saved data anytime from the local computer by using the programs.

From the LabVIEW environment, to record the run-time ECG signals, a .txt file must first be created in the local computer. The file is selected from a dialog box in the window 'Online ECG Signal' of the GUI, as shown in the Figure 5-31. This manipulation accompanies the signal display so that the documented data are 'what you record is what you see'. Clicking the start button  starts recording and pressing the stop button 

is to end the operation. After this, the ECG data within this period are stored in the appointed .txt file.

As long as the ECG data are stored in a local .txt file, recalling them for further analysis is straightforward. To retrieve the data, the user selects the window 'Load ECG' in the GUI and locates the desired file from a dialog box. The user interface in the LabVIEW environment is shown in Figure 5-32.

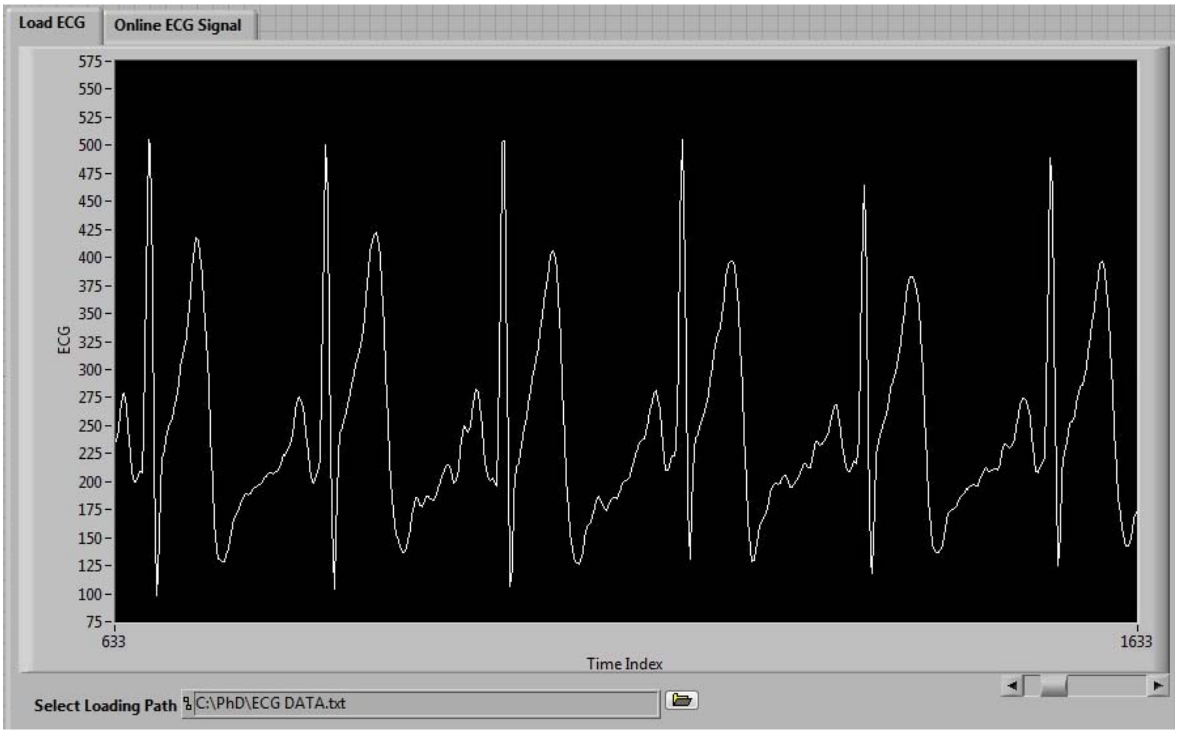


Figure 5-32 Data retrieval user interface in the LabVIEW

Clicking the start button  displays the signal and a direct and instantaneous view of the ECG signals in the window is provided. Moreover, a slide bar offers an inspection of the different segments of the signals in time zone.

To recall the ECG signals from the saved Excel file is simply to plot the data to a chart in the same file. The reappearance of recorded data has the same curve as what have been displayed on the web page.

5.6 Discussion of the web browser solution

The display and retrieval module can be realised using the LabVIEW program as described above. Alternatively, the signals may also be acquired, displayed and stored using a conventional web browser.

Access and display of the real-time signals on a web browser necessitates a dedicated IP address so as to identify the sole object. The IP address is given as ‘192.168.1.118’ in the experiment. If this IP address is given to the web browser on any computer connected to the Internet, the signals will be displayed, as shown in the screenshot of Figure 5-33.

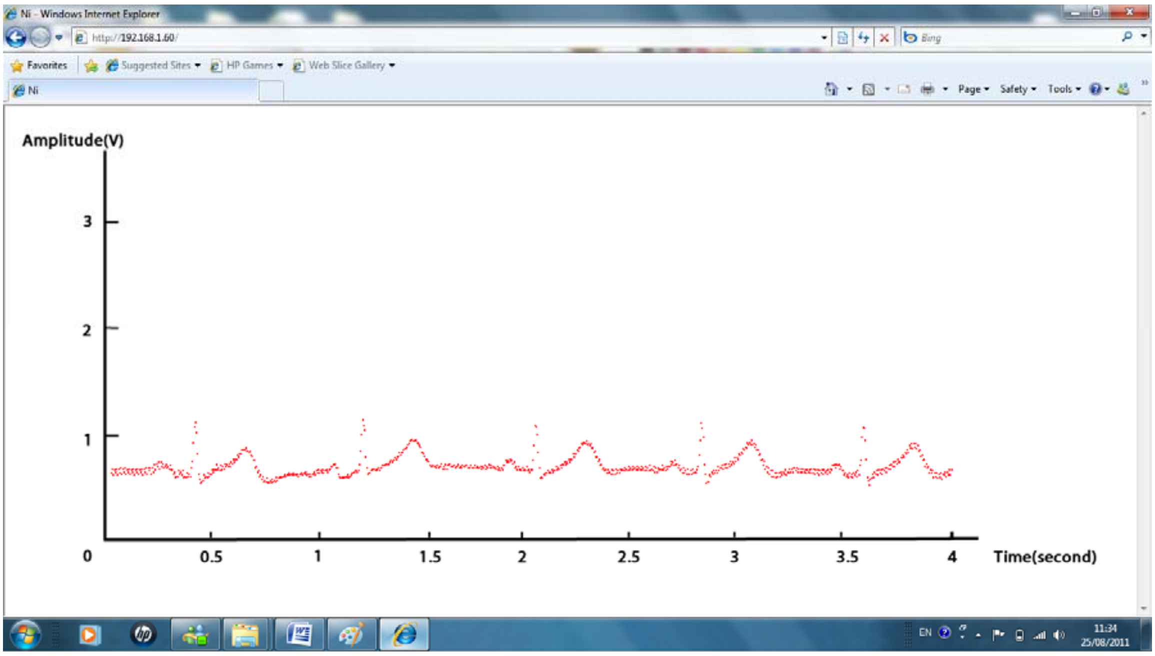


Figure 5-33 Screen view of the real-time ECG signals on the web page

A clock stop was employed to detect how long the signal on the web responded when an event occurred on the signal source. It took less than one second to receive the event response. The real-time figures displayed on the web page are nearly synchronous with the ECG signals from the patient’s side. The quality of the ECG signals as displayed are acceptable for visual monitoring purposes.

A dual channel acquisition system using web browser was also developed. With the specified IP address, Lead I and Lead II ECG signals are displayed synchronously on the same web page, shown in the screen view in Figure 5-34.

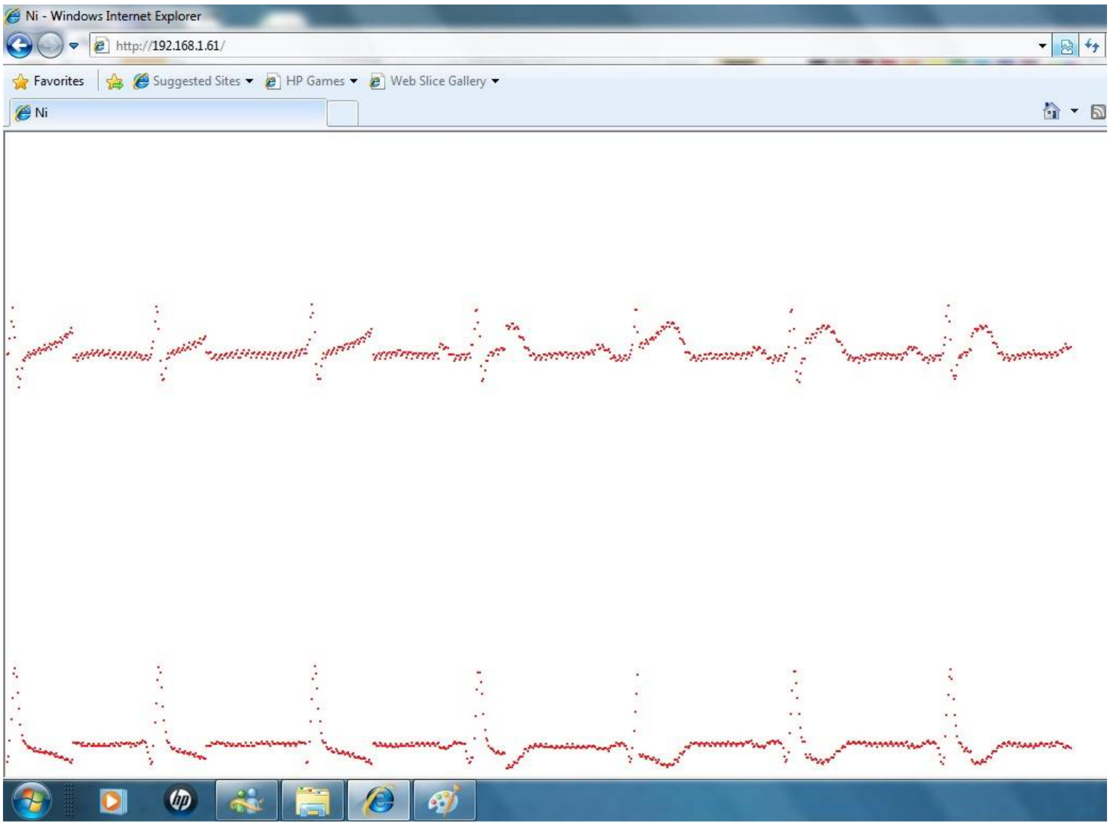


Figure 5-34 Screen view of the dual-channel real-time ECG signals on the web page

The two channel ECG signal acquisition system is demonstrated here successfully. It is worth emphasising that the number of channels that may be transmitted and the quality of the signals are determined by the sample rate and the stream flow rate of the network. More specifically, the sampling rate of the monitored signals dictates the volume of the data that is transferred onto the network. Consequently it has an effect on the channel number, depending on the network data stream speed.

The run-time data may also be stored by the web browser. As shown in the Figure 5-35, a click of the 

button starts importing the real-time ECG data into an Excel file. This Excel file can subsequently be saved to the local disk.

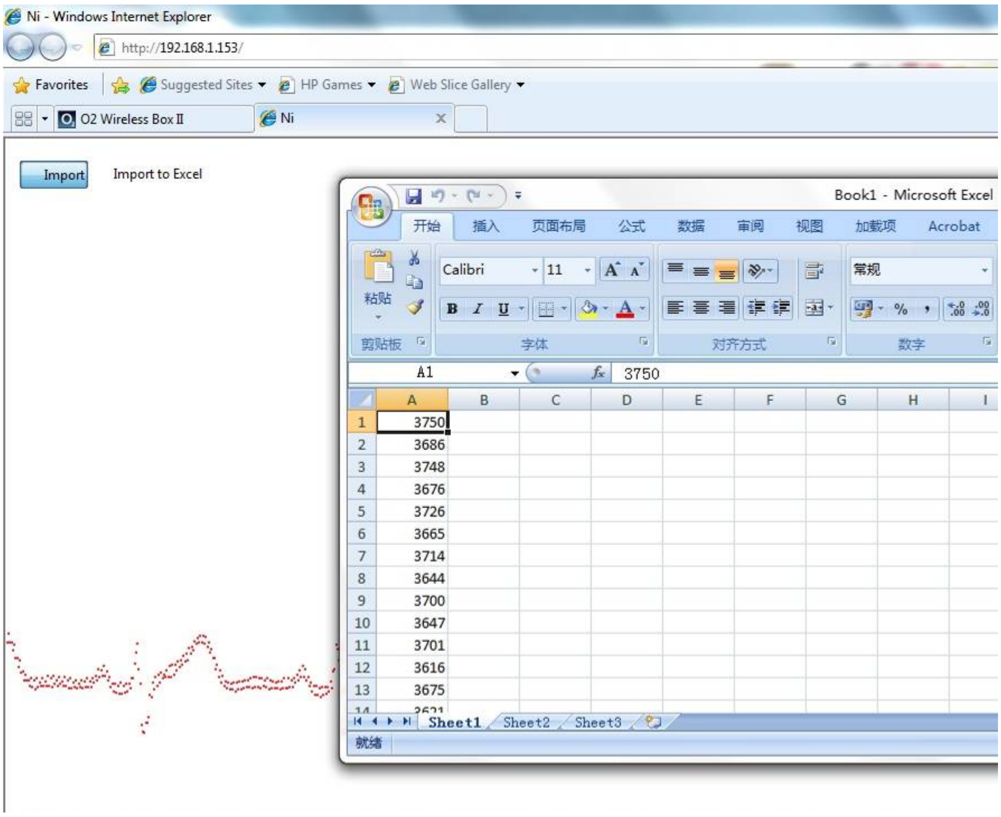


Figure 5-35 Screen view of the run-time ECG data storage on the web page

In the Excel file, each channel is arranged as a single column. In this approach, the real-time ECG signals can be recorded via the Internet without restriction of location and does not require the use of third party software.

In short, this design offers two scenarios to receive, display, store and recall the real-time ECG data on the client site. Both methods have been prototyped and have achieved the objective of signal acquisition, display and storage without data loss. Additionally, advanced digital signal processing and further data applications can be realised in the LabVIEW environment.

5.7 Comparison with other telemedicine system

In contrast to many other telemedicine systems, the system in this work is a truly Internet-based embedded solution. This portable telemedicine instrument provides ubiquitous distributed access to real-time multi-channel biomedical signals by means of Internet transmission. Table 5-8 provides a succinct comparison of this system with other competing technologies.

Table 5-8 Comparison of the designed system and other representative telemedicine systems

System	**Request PC as gateway at the client side**	**Request monitoring centre**	**Wireless connection**	**Real-time data representation**	**Multi-channel**	**Real-time heart rate**	**Body temperature**	**Respiration**
System of this research			✓	✓	✓	✓	✓	✓
Intelligent home care ECG system (D'Angelo et al., 2010)	✓	✓						
Fetal ECG monitoring system (Ibrahimy et al., 2011)	✓	✓						
Portable Multi-life-parameter monitoring system (Qinwu et al., 2011)	✓	✓	✓	✓	✓		✓	
To be continued								

CARA healthcare system (Bingchuan and Herbert, 2011)	✓	✓	✓		✓		✓	
A Zigbee-based telecardiology system (Ernest et al., 2011)	✓	✓	✓	✓				
Pervasive cardiac monitoring system (Hai-ying and Kun-mean, 2010)	✓	✓	✓		✓			
A secure and resource-aware BSN system (Honggang et al., 2010)		✓	✓	✓				
A wireless monitoring system (Rotariu et al., 2011)		✓	✓	✓	✓		✓	✓
Wireless sensor networks (Dilmaghani et al., 2011)		✓	✓	✓				

The system of this research represents multi-sensor real-time biomedical signals, which is more critical for telemedicine application, compared with some off-line only signal manipulation system such as ‘Intelligent home care ECG system’ (D'Angelo et al., 2010) and ‘Fetal ECG monitoring system’ (Ibrahimy et al., 2011). Some research work developed instant alarm monitoring system through the network (Rotariu et al., 2011), which obviously provided limited biomedical information for the clinician/doctor. Other designs applied real-time physiological signals by means of employing either PC or PDA as well as data acquisition device (Qinwu et al., 2011), (Bingchuan and Herbert, 2011). In contrast to those, this research work integrates all functionalities of data acquisition and network gateway within a single embedded platform. Thus, the patient merely needs to carry a true portable instrument for better usability. Furthermore, all existing representative telemedicine systems, described in Table 5-8, employ dedicated server machines. This research takes the leading edge and uses a single device to establish the server deployment immunity.

It is explicit that the designed system in this work only employs a portable device on the patient side to acquire the biomedical signals in either wired or wireless mode. The current proposed telemedicine systems must use computer, laptop, smart phone or PDA without exception. Normally several units have to be deployed on the patient end in those cases. The designed system offers ubiquitous accessibility where the Internet is distributed rather than deployment of a dedicated PC or server machine. Additionally, this system has capability of multiple sensors/channels signals transmission, which is appropriate for biomedical function monitoring. It offers a further compact, robust, wearable solution for the telemedicine system.

5.8 Summary

In brief, a prototype Internet-based hardware and software system for ECG acquisition, transmission, display and storage has been developed and tested. The performance of each sub-module, especially all adopted filters, has been examined. The effect or result of the processed ECG signals in every sub-module has been characterised and analysed. Both single channel and dual-channel manipulation have been depicted. Thus, the potential capability of a multi-channel system can be extended based on the results of the dual-channel implementation. The ECG signals are acquired, manipulated and transmitted onto the Internet by the true embedded modules. In accordance with the specified IP address, the ECG data are received, represented and stored in the client computer by either the web browser or the designed LabVIEW programme. Ultimately, the recorded data can be recalled and re-displayed using the appropriate software.

CHAPTER 6

Conclusions and Further Work

6.1 General conclusions

With reference to most current research in telemedicine systems, almost all the literature suggests the use of cumbersome equipment/computer/laptop/smart phone/PDA as the data transmission gateway from the patient side, and heterogeneous networks to transfer the signals. Consequently, the redundant equipment and the heterogeneous networks make such telemedicine systems awkward and less efficient. On the other hand, the current exploiting scenarios and solutions focus in the face of gathering the assorted different operation systems. The whole system is simply assembled by the different independent systems but not highly ‘integrated’ and ‘compatible’ each other. It is highly possible to result in a degradation of the system’s integration and performance.

The target of the research programme was to design and develop an innovative Internet-based real-time DSP system for telemedicine. The envisaged scenario was to design and implement a truly embedded system and a portable instrument with connectivity to the Internet. There would be no redundant bulky equipment/computer/laptop/smart phone/PDA, but a miniaturised device used by the patient. The clinicians are able to access to the patient’s data and monitor the patient’s clinical condition in real-time via the Internet from any place in the world. The features of the overall system are of high-level integration and compatibility, as well as the broad coverage and wide accessibility by virtue of the Internet.

In this context, the scientific novelty of the Internet-based real-time DSP system in the application of telemedicine has been proven and assessed successfully, as evidenced by the development of the hardware and software prototype. The achievements have been discussed in the previous chapters, and are now recapitulated as follows:

- The feasibility of the multiple channel structure has been proven by the development of two ECG channels in the multiple-sensor module. These two channels carry the Lead I and Lead II ECG signals respectively and ultimately are reproduced almost synchronously on the web page. The transmission speed and the carrying capacity of the networks determine the total delivered data, which is a function of the number of channels and sampling rate.

- The signal conditioning hardware has been designed to amplify the raw signals by a factor of 1000, recover the ECG signals from noise via the multiple filters, and regulate the voltage bias of the signals to a proper level according to the ADC input range. The PCB for the signal conditioning module was produced in compliance with the portable requirements.

- The stand-alone data transmission hardware was implemented successfully, which enables the ECG signals to be propagated via the Internet in real time. This circuit board can be connected to the network either through the Ethernet cable or via the wireless adapter over Wi-Fi. In this way, the system can operate in one of two optional modes: static or mobile mode. The preliminary prototype of the ambulatory solution has been implemented.

- Software applications as either a web browser or a LabVIEW system have been produced to enable the display and storage of the run-time ECG signals on the client side. The dynamic display of the real-time ECG signals in the user interfaces demonstrates that the main purpose of the research programme has been achieved.

- Furthermore, the storage and retrieval functions of the real-time signals have been completed for both the web browser and the LabVIEW environments. The users can view, record and recall the patient's biomedical signals in real-time from any location in the world over the Internet.

In summary, the features of this system include signal collection, processing, visualisation, storage and retrieval, which enables the clinician to monitor a patient globally in real-time over the Internet. The hardware expedites the operation of the miniaturisation of the telemedicine system. The software environments offer multiplicity and feasibility for the applications in varied fields. Although the prototype is yet at a preliminary stage, the generalised concept of the designed system characterises compactness, robustness, reliability and cost-effectiveness.

6.2 Further work

Nevertheless, although the designed system is both inventive and innovative respecting telemedicine research, the end-point in this work has not yet been reached. There is still much potential for improvement towards a commercial product. The further work is suggested and indicated as follows:

- As described in the Chapter 3, the ADC selected for the design is an eight channel device. If numerous biomedical signals need to be acquired in future applications, more ADCs in combination with multiplexers would be required in order to extend the multi-channel capability of the system.

- Based on the multi-channel capability of this system, diverse sensor applications can be applied in the system, for example, multiple leads ECG signals, heart rate, body temperature, respiration signals and so on. The corresponding signal conditioning circuits for the different signals must also be incorporated. However, all the signal conditioning circuits can be restructured or modified in the signal conditioning module only with little alteration of the entire system framework.

- For convenience and practicality, a miniature rechargeable battery is required combined with a power management system. Clearly, for ambulatory purposes a power cable cannot be employed. The rechargeable power supply module must provide a steady, voltage, typically 3.3 V, rated at 500 mV.

- The miniaturisation of the hardware will determine whether or not this telemedicine system is practical and of commercial potential. For that reason, the future electronics should be produced as a single board, highly integrated system. The portable/wearable hardware device should integrate four modules of the system: the signal conditioning module, the data transmission module, the wireless adapter module and the power supply module.

- A friendly user interface can be added to the system towards the evolution of a mature commercial product. A future web-based interface should include time and amplitude scales for the real-time ECG signals as well for the various other biomedical signals. In the environment of the LabVIEW software, the user graphical interface may be designed as a virtual instrument providing full data display and analysis.

- With regard to the utilisation of the textile ECG electrode, even though it is still an immerging technology, this dry type sensor represents an essential component of any future system based on intelligent clothing. With the exploitation of such sensors, the multiple-sensor module can be easily reconfigured to replace the Ag/AgCl electrodes.

- By integrating the textile ECG sensors, intelligent clothing as a wearable measurement platform has been proposed recently. The intelligent clothing is a prospective target for the telemedicine system, which will offer comfortable monitoring combined with minimal impact to the user's daily life. All biomedical sensors and hardware elements may be embedded within the garment.

6.3 Conclusion

A prototype of the hardware and software telemedicine system has been developed and assessed, incorporating real-time DSP and using the Internet as a communication pathway. As such, it attains the objective of the envisioned research proposal. It can be stated that the design scenario is absolutely suitable for a future practical telemedicine system. The flexibility and modularity of the framework enables the system to be scalable. Based on the framework of the designed system, it could have extensive applications in the fields of sensor networks, surveillance, home monitoring, education, industry, medicine as well as entertainment.

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APPENDIX **A**

Section 5.2.3

The peak-to-peak amplitudes of the high-pass filter outputs are recorded here.

Table A-1 Peak-to-peak amplitudes of the high-pass filter outputs

Frequency (Hz)	0.1	0.2	0.3	0.4	0.5	0.6	0.7	0.8	0.9
Amplitude (V)	0.46	0.80	1.08	1.30	1.46	1.58	1.68	1.74	1.80
Normalised value	0.23	0.40	0.54	0.65	0.73	0.79	0.84	0.87	0.90
Frequency (Hz)	1.0	1.1	1.2	1.3	1.5	1.8	2.0	3.0	5.0
Amplitude (V)	1.84	1.88	1.90	1.92	1.96	1.98	2.00	2.00	2.00
Normalised value	0.92	0.94	0.95	0.96	0.98	0.99	1.00	1.00	1.00

Section 5.2.4

The peak-to-peak amplitudes of the notch filter outputs at various different frequencies were recorded in the following table. In addition, the normalised values of these amplitudes are calculated here.

Table A-2 Peak-to-peak amplitudes of the notch filter outputs

Frequency (Hz)	1	5	10	15	20	25	30	35	40
Amplitude (V)	1.96	1.90	1.56	1.28	0.94	0.728	0.552	0.408	0.288
Normalised value	0.98	0.95	0.78	0.64	0.47	0.364	0.276	0.204	0.144
Frequency (Hz)	45	50	53	58	65	70	80	100	130
Amplitude (V)	0.188	0.112	0.088	0.128	0.224	0.288	0.416	0.624	0.856
Normalised value	0.094	0.056	0.044	0.064	0.112	0.144	0.208	0.312	0.428
Frequency (Hz)	160	200	250	300	350	400	500	600	700
Amplitude (V)	1.05	1.24	1.40	1.51	1.68	1.72	1.80	1.88	1.88
Normalised value	0.525	0.62	0.7	0.752	0.84	0.86	0.9	0.94	0.94

Section 5.2.6

The numerical values of the peak-to-peak amplitudes of the processed signals are recorded in accordance to the frequency serial.

Table A-3 Peak-to-peak amplitudes of the low-pass filter outputs

Frequency (Hz)	10	50	80	90	95	100	105	110	113
Amplitude (V)	2	1.96	1.80	1.68	1.56	1.44	1.32	1.20	1.12
Normalised value	1	0.98	0.90	0.84	0.78	0.72	0.66	0.60	0.56
Frequency (Hz)	116	120	125	130	135	140	150	160	180
Amplitude (V)	1.04	0.96	0.84	0.76	0.68	0.60	0.52	0.44	0.32
Normalised value	0.52	0.48	0.42	0.38	0.34	0.30	0.26	0.22	0.16

APPENDIX B

Table B-1 The measured length changes of the strap and their corresponding voltage outputs

Test	Elongations (mm)	Voltage outputs (mV) Group 1	Voltage outputs (mV) Group 2	Voltage outputs (mV) Group 3	Voltage outputs (mV) Group 4	Voltage outputs (mV) Group 5
1	0	*12*	*12*	*13*	*13*	*13*
2	5	*61*	*56*	*72*	*65*	*61*
3	10	*194*	*174*	*181*	*198*	*178*
4	15	*316*	*269*	*283*	*305*	*262*
5	20	*450*	*422*	*402*	*421*	*389*
6	25	*581*	*550*	*562*	*572*	*562*

APPENDIX C

Table C-1 A group of duplicate measurements for the temperature sensor evaluation

Test1	
Reading by the sensor (˚C)	Reading by the thermometer (˚C)
43.4	43.6
42	42
40	40.3
38.2	38.6
36.6	37
35.1	35.9
34	34.7
32.6	33.3
31	31.6
Test2	
Reading by the sensor (˚C)	Reading by the thermometer (˚C)
43.2	43.5
41.6	41.5
39.8	40.1
37.8	38.1
36.1	36.7

34.8	35.6
33.3	34.5
32	32.9
30.3	31
Test3	
Reading by the sensor (˚C)	Reading by the thermometer (˚C)
42.4	43
41.2	41.9
39.1	39.8
37	37.5
35.5	36
34.1	34.4
32.6	33.6
31.3	32.1
29.8	30.6
Test4	
Reading by the sensor (˚C)	Reading by the thermometer (˚C)
43	42.8
41.3	41.1
39.5	39.6
37.2	37.1
35.6	35.8

34.2	34.2
32.8	33.1
31.8	32.4
28.9	30.5
Test5	
Reading by the sensor (˚C)	Reading by the thermometer (˚C)
43.2	43.1
41.9	41.8
40.2	40.5
38	37.7
35.5	36.2
34.4	35
33	33.8
32	33
30	31.1

APPENDIX D

Table D-1 A series of body temperature records through the daytime

Time	Multimeter reading (mV)	Measured temperature (˚C)	Calibrated value (˚C)
9:00	856	35.6	36.4
9:30	858	35.8	36.6
10:00	858	35.8	36.6
10:30	860	36.0	36.8
11:00	860	36.0	36.8
11:30	860	36.0	36.8
12:00	859	35.9	36.7
12:30	859	35.9	36.7
13:00	861	36.1	36.9
13:30	861	36.1	36.9
14:00	861	36.1	36.9
14:30	862	36.2	37.0
15:00	862	36.2	37.0
15:30	862	36.2	37.0
16:00	862	36.2	37.0
16:30	861	36.1	36.9
17:00	861	36.1	36.9

17:30	860	36.0	36.8
18:00	859	35.9	36.7
18:30	858	35.8	36.6
19:00	858	35.8	36.6
19:30	860	36.0	36.8
20:00	860	36.0	36.8
20:30	859	35.9	36.7
21:00	859	35.9	36.7

APPENDIX E

Table E-1 System requirements for NI LabVIEW development systems

Development Environment	Windows
Processor	Pentium 4/M or equivalent
RAM	1 GB
Screen Resolution	1024 x 768 pixels
Operating System	Windows 7/Vista/ (32- and 64-bit) Windows Server 2008 (64-bit) Windows Server 2003/XP SP2 (32-bit)
Disk Space	3.3 GB (includes default drivers from NI Device Drivers DVD)

APPENDIX F

Ethics approval form

MANCHESTER
1824

The University of Manchester

Secretary to Research Ethics Committees
Room 2.004 John Owens Building

Tel: 0161 275 2206/2046
Fax: 0161 275 5697
Email: timothy.stibbs@manchester.ac.uk

Compliance and Risk Office
University of Manchester
Oxford Road
Manchester, M13 9PL

ref: ethics/12114

Professor Patrick Gaydecki,
School of Electrical and Electronic Engineering,
Sackville Street Building, F45

20th June 2012

Dear Patrick,

Research Ethics Committee 3
Zhu, Gaydecki: An internet-based real-time DSP system design in the application of telemedicine (ref 12114)

I write to confirm that the above project has been reviewed and given a favourable ethical opinion.

This approval is effective for a period of five years and if the project continues beyond that period it must be submitted for review. It is the Committee's practice to warn investigators that they should not depart from the agreed protocol without seeking the approval of the Committee, as any significant deviation could invalidate the insurance arrangements and constitute research misconduct. We also ask that any information sheet should carry a University logo or other indication of where it came from, and that, in accordance with University policy, any data carrying personal identifiers must be encrypted when not held on a university computer or kept as a hard copy in a location which is accessible only to those involved with the research.

Finally, I would be grateful if you could complete and return the attached form at the end of the project or by June 2013.

Yours sincerely

Timothy Stibbs

Dr T P C Stibbs
Secretary to the University Research Ethics Committee

Enclosed: Report form

The University of Manchester, Oxford Road, Manchester M13 9PL Royal Charter Number: RC000797

Document E

The University of Manchester

MANCHESTER 1824

APPENDIX B

Project Title

CONSENT FORM

If you are happy to participate please complete and sign the consent form below

	Please Initial Box
1. I confirm that I have read the attached information sheet on the above project and have had the opportunity to consider the information and ask questions and had these answered satisfactorily.	✓
2. I understand that my participation in the study is voluntary and that I am free to withdraw at any time without giving a reason and without detriment to any treatment/service	✓
3. I understand that the interviews will be audio-recorded	✓
4. I agree to the use of anonymous quotes	✓
5. I agree to my GP being informed of my participation in the study	✓
6. I agree that any data collected may be passed to other researchers	✓

I agree to take part in the above project

Name of participant	Date	Signature
Ni Zhu	22/07/2012	Zhu Ni

Name of person taking consent	Date	Signature
P. GRAYDECKI	22/07/2012	[illegible]