IMPLEMENTATION OF INTERACTIVE REMOTE PHYSIOLOGICAL MONITORING AND FEEDBACK TRAINING SYSTEM

A Thesis
Presented to
The Graduate Faculty of the University of Akron

In Partial Fulfillment
of the Requirements for the Degree
Master of Science

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December, 2006
IMPLEMENTATION OF INTERACTIVE REMOTE PHYSIOLOGICAL MONITORING AND FEEDBACK TRAINING SYSTEM

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Thesis

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ABSTRACT

Remote Physiological Monitoring as an application of telemedicine holds great promise for providing efficient and economical health care. Most telemedicine applications have been limited to the use of interactive video for physician consultations, educational, research and administrative purposes. In this research project, an attempt was made to design and implement software that provided an interactive remote physiological monitoring system with instantaneous feedback to the physician at any remote location, about the general health of the patient, through the Internet. The software acquired sixteen channels of raw physiological signals and provided twenty parameters, which included heart rate, blood pressure, temperature and respiratory rate, describing the physiological condition of the patient. Reliable algorithms were implemented for the processing of raw physiological signals to provide accurate values of useful physiological parameters. The remote communication between the physician end and the patient end was achieved using the Transmission Control Protocol/Internet Protocol (TCP/IP). The data acquisition, signal processing and display of the processed output at patient’s site were controlled by the physician at the remote site. The design of software was such that, it was interchangeably used for patient monitoring as well as training of subjects to control their physiological responses to various environmental stressors. The software was designed and implemented in the Microsoft Visual Basic 6.0® Integrated Development Environment.
ACKNOWLEDGEMENT

It has indeed been an honor, privilege and pleasure to work with Dr. Bruce C. Taylor. The discussions I have had with him, technical and non technical, have always inspired me to think differently. I consider his encouragement, guidance and support, throughout the duration of my research and my study here, as invaluable in developing my attitude towards science and engineering. I would like to express my gratitude to Dr. Dale H. Mugler and Dr. Daniel B. Sheffer for their advice and suggestions.

I take this opportunity to gratefully acknowledge the grant from NASA Ames Research Center for this project. I would like to thank Dr. Patricia Cowings and Dr. William Toscano, for their support and valuable feedback regarding the functioning of the software, without which the research would not have been possible.

I would like to make a special mention of Mr. Rick Nemer, who was ever forthcoming with technical help and guidance during the course of the project, and Ms. Bonnie Hinds.

This project would not have been possible without the support and guidance of my friend, philosopher and guide, Dr. Soumyadipta Acharya. I would also like to thank my parents for showing faith in me and all my friends for their support.

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CHAPTER I

INTRODUCTION

Physiological monitoring is an essential part of medical care and has improved significantly over the years. Clinical monitoring systems have been developed in response to the need to acquire patient physiological data and have only recently been accepted as standard equipment in the operating room. One of the reasons for the improvement in health care over the last 50 years has been due to this acceptance and also the advances made in the monitoring systems for patients. Instruments, which measure various physiological parameters and give an idea of the general health of the patient, have been developed. Similar monitoring systems have also been used in feedback training of personnel who work in conditions of physical and psychological stress. This kind of feedback training helps train the trainees to control physiological responses to stressors in the work environment.

Any physiological monitoring system design is based on the premise that physiological events can be converted to electrical signals. Special purpose processors or software, running on special computers, are incorporated to perform specific computational tasks on these signals to detect physiological events or measure a physiological parameter. Most clinical monitoring systems aim at acquiring the four
traditional vital signs: temperature, pulse or heart rate, respiratory rate and blood pressure. Effective signal processing is an essential part of these physiological monitoring systems. The advent of digital signal processing techniques has made it possible to design and implement signal processing software. The advantages that the software techniques of signal processing carry over conventional methods, using hardware, are that they are cost effective, easily modifiable and easily implemented. These techniques give an acceptably close approximation of outputs obtained from conventional methods of signal processing. Any error introduced can be overcome by making use of various numerical error correction techniques or techniques of adaptive filtering.

1.1 History of monitoring systems

Physiological monitoring has been an integral part of medical diagnosis since antiquity. Ancient Greek physicians, including Hippocrates (ca. 460-377 B.C) and Galen (ca. 129-200 A.D.) had palpated the pulse and described its variations [29]. Herophilos emphasized the importance of heart rate in diagnosis, who is reported to have transported water clocks to the patient and quantitate pulse in patients with fever [29]. The earliest monitoring system for human patients was developed in the 1940’s by a neurosurgeon, Wilder Penfield, and his associates at the Montreal Neurological Institute. He developed a sophisticated arrangement for the measurement electroencephalograph signals and diagnosis or regions in the brain where lesions existed [29]. Penfield had also developed another multi-channel recording system to serve anesthesiologists [29]. The late E.A. Pask (ca. 1948), is known to have developed a multi-channel recording system, to record
ECG, direct blood pressure and respiration [28]. This equipment was used in the recovery room following surgery. Its use was abandoned soon after the end of Pask’s sabbatical at the Institute from Newcastle-on-Tyne, England. [29]

With the development of multi-channel cathode ray tube displays and development of smaller electronic equipment in the 1960’s, the development of monitors of physiological events received a great impetus. In spite of initial resistance to such automation among ward personnel, a concerted educational program, the development of the concept of intensive care units and the urgent need to monitor patients undergoing open heart and vascular replacement surgery, did much to speed the acceptance of such systems. Recent advances in computer technology, the recognition of the advantages of digital signal processing systems over analog systems and development of better algorithms for physiological data processing, has led to the realization of more reliable data acquisition and monitoring systems. These systems have become indispensable in today’s hospital environment.

1.2 Telemedicine and physiological monitoring

Defined broadly, telemedicine is the use of electronic information and communications technologies to provide and support health care when distance separates the participants. The term is also applied more narrowly to medical applications that use interactive video typically for specialty or subspecialty physician consultations [5]. Sometimes the term telehealth is used to encompass educational, research and
administrative uses as well as clinical applications that involve nurses, psychologists, administrators and other non-physicians.

With the advent of the Internet and development digital communication technology, high rates of data exchange has become possible. Transmission Control Protocol / Internet Protocol (TCP/IP) is the most widely used protocol for transmission of data over the Internet. The protocol guarantees reliable and in-order delivery of sender to receiver data. The use of the Internet in remote monitoring holds great promise as it reduces the cost of dedicated transmission lines as was the case in the past and also eliminates the need of satellites for transmission of data which means high initial cost of capital for any company providing such a facility.

1.3 Overview of AFTE

As mentioned previously, monitoring systems are also used in feedback training of personnel exposed to stressful conditions in their work environment, e.g. spacecraft crew and fighter pilots. One such feedback training technique, Autogenic Feedback Training Exercise (AFTE) is an innovative technology developed by NASA to help its astronaut’s combat unwanted physiological responses to external stimuli by controlling their Autonomic Nervous System (ANS), during space flight. The ANS is responsible for controlling and regulating involuntary bodily functions, such as breathing, heartbeat, sweating, blood vessel dilation, skin conductance, muscle reactivity and glandular secretions. AFTE teaches people to control up to twenty such physiological responses. The success of such an exercise requires a highly efficient data acquisition and signal
processing system, which empowers the trainer to observe the physiological and psychological state of the subject and allows the subject to take stock of his own performance through the training sessions of AFTE.

1.4 Objective of the study

This research study was aimed at examining the possibility of implementing a comprehensive remote physiological monitoring and feedback training system that would efficiently provide information about the general health condition of the patient or subject, control of the data acquisition and processing to adjust to the needs of individual patients or subjects and the same would be possible from either onsite or from a distant location.

The specific objectives of this research project were:

• To develop efficient and robust signal processing software that processes sixteen channels of raw physiological data and effectively extracts twenty parameters, which are indices of health of the patient or subject; to be used for monitoring and/or feedback training.

• To develop a communication system which would enable the physician or training expert at the remote site to control the data acquisition, processing, display and storage of processed data from any distant location where the Internet is available.
1.5 Hypothesis

Null Hypothesis

The remotely controlled monitoring system that has been developed can be used to accurately monitor the twenty physiological indices that are determined from the sixteen channels of raw physiological data given as input to the system. The means of the 20 indices are not different from the means that can be calculated by examining the raw physiological waveforms, i.e. $\mu_1 = \mu_2$.

Alternate Hypothesis

The remotely controlled monitoring system that has been developed cannot be used to accurately monitor the 20 physiological indices that are determined from the sixteen channels of raw physiological data given as input to the system. The means of the 20 indices are different from the means that can be calculated by examining the raw physiological waveforms, i.e. $\mu_1 \neq \mu_2$. 
A comprehensive and reliable signal processing system was the desired outcome of this research study and in order to achieve that, the system made use of signal processing algorithms that not only displayed the correct values but also updated at an optimum rate, so that small changes were registered and displayed, especially in the case of high frequency signals. Careful formulation of the communication protocol and the data storage was also required to allow the software to run without computational overhead. The following sections list the literature that was reviewed for deciding on the algorithms and methods that were best suited for achieving the desired results.

2.1 Stroke volume, cardiac output and thoracic impedance

Cardiac output is defined as the volume of blood ejected by the heart per unit of time [8]. Its accurate determination provides information that could be useful in the treatment of a critically ill patient. Geddes [1] translated Adolf Fick’s original paper on the development of a method to measure cardiac output by Fick’s method, which remains the gold standard with which the methods of cardiac output measurements are compared. Though quite accurate and reliable, the use of Fick’s method in the clinical environment has declined since it requires steady state conditions and extraction of blood samples.
One non-invasive method of measuring cardiac output uses the impedance cardiograph signal for the measurement of the stroke volume and the cardiac output is deduced from it. Underwood and Gowing (1965) reported that the impedance between widely spaced electrodes on experimental elements reflected blood volume [1]. In their experiments on dogs and cats, they observed that a 10% change in blood volume produced a 1.5Ω change in impedance. Geddes and Baker verified these observations, using a tetra-polar electrode arrangement applied to dogs. Rappoport and Ray (1927) recorded that impedance changed by 10% with each beat; in a tortoise heart as it was kept beating in vitro [1]. Rushmer et al. (1953), in their study on the right and left ventricles of dogs, recorded an increase in impedance during systole and a decrease in impedance during diastole [1]. Mello-Sobrinho (1963) reported in their study using canine hearts, that a 5Ω change was equivalent to 1-ml change in volume [1]. Cardiac chambers were found to function as conductivity cells of varying dimensions [1]. Palmer (1970) obtained calibration factors of 0.5-1.7 Ω/ml [1] by measuring cardiac output. Atzler and Lehmann (1932) and Atzler (1933, 1935), placed metal electrodes in front and behind the thorax and detected ultra high frequency impedance changes synchronous with ultra cardiac activity. Their method was mainly capacitive and was called “Dielektrographie”. [1]

The correction method developed by Kubicek et al. (1966) assumes that at the beginning of inflow to the inter electrode segment, outflow is minimal and inflow is maximal. Therefore, the extrapolation of the steepest part of the impedance pulse, if continued to the end of the ejection period T, which is often taken as the dichrotic notch in the impedance record would provide an impedance change ΔZ that is corrected for
outflow [1]. The calculation of the stroke volume from the value of $\Delta Z$ is illustrated in Figures 2.1 and 2.2

\[
SV = \rho \cdot \left(\frac{L}{Z_0}\right)^2 \cdot T \cdot \left(\frac{dZ}{dt}\right)_{\text{max}}
\]  \hspace{1cm} (2.1)

where,

$\rho$ = the resistivity of blood (\(\Omega\)-cm)
L = the separation (cm) between the two inner (2, 3) potential measuring electrodes,

\[ Z_0 = \text{the basal impedance between potential measuring electrodes (2, 3)}, \]

\[ (dZ/dt)_{\text{max}} = \text{the maximum rate of change of impedance (Ω/sec)}, \]

\[ T = \text{Ventricular ejection time} \]

2.2 Respiratory rate

Breathing or pulmonary ventilation is accomplished by expansion and contraction of the lungs. This is achieved (1) by upward and downward movement of the diaphragm and (2) by elevation and depression of the ribs to increase and decrease the antero-posterior of the chest cavity. The rate of this expansion and contraction of the lungs is referred to as respiratory rate. Respiratory rate is included among the four traditional vital signs [29], which most often has to be monitored during spontaneous breathing as well as during ventilatory support [Roback and Nelson et al.]. Commonly used respiratory rate monitors are based on trans-thoracic impedance (TTI) [24][25], spirometry, electromyography [26] or respiratory inductance plethysmography [27].

2.3 Heart rate and QRS detection

When the cardiac impulse passes through the heart, electrical current also spreads from the heart into the adjacent tissues surrounding the heart, a small proportion of which spreads all the way to the surface of the body. Electrical potentials generated by this current can be measured by placing electrodes on either side of the heart. The waveform
representing these potentials is called the electrocardiograph (ECG) signal. The measurement of potential can be carried out by placing electrodes on two limbs on either side of the heart (typically on the shoulders or wrists). This arrangement is referred to as the standard bipolar limb lead arrangement. The electrodes in this arrangement form a triangle referred to as the Einthoven’s triangle. There are three recording limbs in this triangle and they are referred to as leads I, II, III. A typical ECG obtained across lead II is shown in Figure 2.3. A normal ECG consists of a P wave, caused by atrial depolarization before atrial contraction begins; a QRS complex caused by ventricular depolarization before their contraction and a T wave caused by the ventricular repolarization [22].

![Figure 2.3: Electrocardiogram](image)

The efficient and accurate detection of the QRS complex is critical for the successful implementation of this project. The reciprocal of the elapsed time between two successive QRS complexes multiplied with 60 would give the instantaneous heart rate. Also, the QRS complex, when detected accurately, can be used as a gating signal for the efficient processing of other signals like blood pressure and pulse volumes.
The power spectrum of the ECG can provide information about the frequency spectrum of the QRS complex. It has been reported by Afonso [2] that the power spectrum (based on FFT) of a set of 512 sample points that contain approximately two heartbeats results in a series of coefficients with a maximal value near a frequency corresponding to the heart rate. The ECG waveform, in addition to the QRS complex, contains P and T waves, 60 Hz noise from power line interference, EMG from muscle, motion artifact from the electrode and skin interference [2]. According to studies of spectral plots of ECG and the QRS complex from 3875 beats by Thakor et al., in 1984, a maximum SNR value is obtained for a band pass filter with a center frequency of 17 Hz and Q of 3 [2].

In many QRS detection algorithms, high and low pass filtering is carried out separately. The filtered signals are then used for the generation of a feature signal in which the occurrence of a QRS complex is detected by comparing the feature against fixed or adaptive thresholds. The high-pass filter is often implemented as a differentiator [11].
A real-time QRS detection algorithm was developed by Pan and Tompkins (1985) [13] and was further described by Hamilton and Tompkins (1986). Pan and Tompkins have claimed in their paper [13] that 99.3 % of the QRS complexes were correctly detected, using the algorithm proposed by them, for the standard 24 hour MIT/BIH arrhythmia database. Their method recognizes QRS complexes based on analyses of the slope, amplitude and width [2]. Software QRS detectors typically include one or more of the 3 different types of processing steps: linear digital filtering, non linear transformation, and decision rule algorithm [12]. Linear processes include a band-pass filter, a derivative and a moving window integrator. The nonlinear transformation that is used is signal amplitude squaring. Adaptive thresholds and T-wave discrimination techniques provide part of the decision rule algorithm [13]. Pan and Tompkins used an ECG signal sampled at 200 samples/s. They implemented a band-pass filter which eliminated the 60 HZ interference, muscle noise, baseline wander, and T-wave interference. The desirable pass-band to maximize the QRS energy is approximately 5-15 Hz. The desired band-pass filter was implemented using a cascaded low-pass and high-pass filter. After filtering, the
signal was differentiated to provide the QRS complex slope information. In the method suggested by Pan and Tompkins, a five point derivative was used. After differentiation, the signal was squared point by point. Moving window integration was used to obtain waveform feature information in addition to the slope of the R wave. According to Pan and Tompkins [13], the width of the window should be approximately the same as the widest QRS complex and was determined experimentally. The QRS complex corresponded to the rising edge of the integration waveform [13]. The transfer functions and the difference equations of the digital filters that have been used in the Pan Tompkins algorithm are listed below. In the equations below, \( H(z) \) represents the transfer functions, \( y(nT-kT) \) represents the output data points and \( x(nT-kT) \) represents the input data points, \( n \) and \( k \) are integers.

Low-Pass Filter:

\[
H(z) = \frac{(1 - z^{-6})^2}{(1 - z^{-1})^2} \quad (2.2)
\]

\[
\Rightarrow y(nT) = 2y(nT - T) - y(nT - 2T) + x(nT) - 2x(nT - 6T) + x(nT - 12T) \quad (2.3)
\]

High-Pass Filter:

\[
H(z) = \frac{(-1 + 32z^{-16} + z^{-32})}{(1 + z^{-1})} \quad (2.4)
\]

\[
\Rightarrow y(nT) = 32x(nT - 16T) - [y(nT - T) + x(nT) - x(nT - 32T)] \quad (2.5)
\]

Derivative:

\[
H(z) = (1/8T)(-z^{-2} - 2z^{-1} + 2z^1 + z^2) \quad (2.6)
\]

\[
\Rightarrow y(nT) = (1/8T)[-x(nT - 2T) - 2x(nT - T) + 2x(nT + T) + x(nT + 2T)] \quad (2.7)
\]
Squaring Function:

\[ y(nT) = [x(nT)]^2 \]  
(2.8)

The thresholds are automatically adjusted to float over the noise. The low thresholds are due to the improvement of the signal to noise ratio by the band-pass filter. Two thresholds are used in the detection in the Pan Tompkins algorithm. The higher of the two thresholds in each of the two sets is used for the first analysis of the signal. The lower threshold is used if no QRS is detected in a certain time interval so that a search back technique is necessary to look back in time for the QRS complex.

2.4 Skin temperature

Body temperature is a reliable and often used method for diagnosis of diseased conditions. It not only tells us about the physical condition of the person but many a time is an indicator of the psychological state of the person.

2.5 Skin conductance level

Skin conductance measurement has been found to be a good method to monitor the autonomic nervous system response. Monitoring of the skin conductance levels can be of immense use in treating patients for various autonomic nervous system disorders. It has also been found to have great role in training pilots and astronauts for autogenic feedback training exercise, developed by NASA to help their astronauts and pilots cope
with stress related physiological responses during flight such as nausea, motion sickness etc. [15].

2.6 Blood volume pulse

Information on the blood circulation in the extremities has been used to treat symptoms such as nausea, migraines and cluster head aches [16] [17][18]. The blood volume pulse can be measured from a photo-plethysmograph signal.

2.7 Surface electromyogram

When a muscle unit contracts, it conducts electrical signals, which can be measured using electrodes placed on the skin above the muscle. The muscle motor units are activated several times a minute by the central nervous system. The amplitude of these electrical potentials can serve as an index of muscle activity. Clancy, Bouchard and Rancourt described a method to estimate the EMG amplitude during dynamic contraction [31]. In their study, “Detection of onset of and termination of muscle activity in surface electromyograms”, Abbinl, Van der Bilt and Van der Glas, used rectified EMG signals to determine muscle activity [32].

2.8 Vagal tone and heart rate variability (HRV)

“Over the last 25 years, HRV analysis has established itself as a non-invasive research and clinical tool for indirectly investigating both cardiac and autonomic system function in both health and disease.” [19]
Heart rate is one of the basic parameters in medical diagnostics. Precise and reliable assessment of its variability still remains an engineering and physiological challenge [7]. Vagal tone or heart rate variability (HRV) is one such method of the assessment of changes that occur in the heart rate. Spectral analysis helps identify the various components of heart rate variability. They are related to different cardiac control activities such as blood pressure, thermal regulation and respiration [7]. The frequency analysis of heart rate has been found to show two spectral peaks, corresponding to a low frequency heart rate variability (0.04-0.15 Hz, related to baroreflexes, representing sympathetic and parasympathetic activity) and a high frequency peak (0.15-0.4 Hz, related to respiration and representing parasympathetic activity) [30]. McCabe, Yongue, Porges and Ackles (1984) have stated that the magnitude of the HRV within the spectral range of respiration can be used as a measure of the strength of parasympathetic activity of autonomic nervous system [9]. The instantaneous heart rate signal is a staircase signal with possible discontinuities at each heart beat. A low-pass filter event series is computed by

\[
f(t) = \sum f(t_k) \frac{\sin(2\pi f_c(t-t_k))}{\pi f_c(t-t_k)}
\]  

(2.9)

This is essentially an ideal low pass filtering of the impulse series with cut off frequency \(f_c\). As stated by Nagel, Han, Hurwitz and Schneiderman [7], the mean frequency of these observations is the actual mean heart rate. It follows that, according to
the Nyquist Criterion, the maximum observable frequency is limited to half the mean heart rate, though in reality, maximum frequency is lower. With a mean heart rate of 60 beats per minute (1 Hz), the bandwidth of the heart rate variability signal is always less than 0.5 Hz. Therefore cut off frequency $f_c$ maybe chosen as 0.5 Hz [7]. Several researchers have reported that phasic changes in respiration have a gating influence on cardiac vagal efferent activity such that during the expiratory phase of respiration a slowing of heart rate is observed. The oscillatory influence of respiration on heart rate has been referred to as Respiratory Sinus Arrhythmia (RSA) [7]. The magnitude of Heart Rate Variability within the spectral range of respiration has been used as a measure of the strength of parasympathetic activity of the autonomic nervous system [7] [9]. Since RSA is primarily mediated through the vagus nerve, the measurement of RSA can be used as index of cardiac vagal tone [10]. Yongue, McCare, Porges, Rivera, Kelley and Ackles quantified the component of HRV associated with respiration, in rats, which is proposed to be sensitive to vagal influences on heart, using the spectral analysis [10]. In the method adopted by them, the inter-beat interval was converted into time based sequence of successive 100 millisecond windows. A 15 point polynomial was moved stepwise through the heart rate and respiratory data of each trial. The polynomial procedure functioned as a filter with frequency cut-off 0.67 Hz. For this experiment, the band of frequencies from 0.8 to 2 Hz was found to encompass the peak respiratory frequency for all trials, and was used as the band of respiratory frequencies for analysis.
2.9 Telemedicine system

One of the earliest instances of telemedicine was the transmission of electrocardiogram signals over telephone lines by Einthoven as early as 1905 [6]. Both interactive and non interactive technologies are increasingly used or promoted for remote health monitoring of health status in homes and other settings [5]. Physiological parameters such as blood pressure, heart rate, blood glucose level and pulse oximetry data can now be reliably measured and transmitted, using monitoring devices connected to specialized modems [5]. Razadeh and Evans [6] described a system which could transmit up to six low frequency data channels in the presence of a full duplex speed channel using one telephone line.

With the advent of the Internet, it has become possible to interconnect various disparate physical networks and make them function as a coordinated unit [14]. the Internet has expanded rapidly and so have its applications. The TCP/IP is the underlying communication protocol for most of the communication that takes place on the Internet. It contains rules that contain details of message formats, describes how a computer responds when a message arrives and specifies how a computer handles errors or other abnormal conditions. The predominant usage of the TCP/IP protocol for data communication over the Internet and the ease of implementation of a TCP/IP based system of communication make it the ideal choice as the protocol for the communication in the Remote Physiological Monitoring and Feedback Training software.
CHAPTER III
METHODOLOGY

This chapter describes the algorithms, the communication system and graphical user interface (GUI) that have been implemented in the monitoring system software. The monitoring system software was developed using Microsoft Visual Basic 6.0. This project can be broadly divided into the following segments.

- Data Acquisition and Signal Processing module.
- Communication and Synchronization module
- Graphical User Interface Development module.

3.1 Development of patient/subject simulator

Any real time processing requires an inflow of a continuous stream of data that would be acquired and processed. It was realized that it would not be feasible to have a human subject from whom data could be acquired every time the system was to be tested and evaluated. Therefore a computer program was written in Turbo C++ which read a data file and provided sixteen channels of data as output, through a digital-to-analog (D/A) converter at a required steady rate [Appendix A]. The Industrial Computer Source D/A converter AOBx/12 card, which makes use of the 12 bit D/A converter AD7237 integrated circuit chip, was used for reading the digital data from the file.
3.2 Data acquisition

Fifteen channels of raw physiological data were acquired at a sampling rate of 250 samples second per channel [3]. It was made possible for the sampling rate to be changed if the user of the software desired to do the same. The WinDaq® ActiveX Control was used to acquire data for processing in the Microsoft Visual Basic 6.0 environment. The information on the hardware used for data acquisition is provided in Appendix B. The channels that were acquired are listed below. Figure 3.1 shows the WinDaq display the channels of data being acquired.

- Channel 1: Left Finger Pulse Volume
- Channel 2: Right Finger Pulse Volume
- Channel 3: Respiratory Signal
- Channel 4: Electrocardiogram
- Channel 5: Temperature
- Channel 6: Left Toe Pulse Volume
- Channel 7: Right Toe Pulse Volume
- Channel 8: Skin Conductance Level
- Channel 9: Left Arm EMG
- Channel 10: Right Arm EMG
- Channel 11: Left Leg EMG
- Channel 12: Right Leg EMG
- Channel 13: Blood Pressure Signal
- Channel 14: Not used
- Channel 15: dZ/dt
- Channel 16: Z₀

Figure 3.1: Display of 16 channels of input data
Description of the WinDaq® Data Acquisition System

The data acquisition was done using the WinDaq DI-720 Data acquisition Module manufactured by DATAQ Instruments. The DI-720 series of instruments uses a 16-bit A/D converter for high resolution measurement accuracy but the maximum accuracy that the WINDAQ software was designed to achieve was 14 bits [DI-720/DI-730/DI-740 Series User Manual]. The DI-720 features software programmable input measurement ranges of ±10V full scale at a gain of 1; ±5V full scale at a gain of 2; ±2.5V full scale at a gain of 4; and ±1.25V full scale at a gain of 8 [21]. In this research, a full scale range of ±10V at a gain of 1 was used for acquiring the data.

3.3 Signal processing

For the intensive signal processing and data communication capability that was required for the implementation of the software, a Dell® OPTIPLEX GX270 computer system, with a 3.4 GHz Intel Pentium® 4 processor and a Random Access Memory of 1GB was used. The digital filters used for the implementation of the signal processing algorithms were designed using the Filter Design and Analysis Tool (FDA Tool) in the MATLAB® 6.1 Toolbox.

The 20 output parameters that were to be calculated were

- Skin Conductance Level
- Temperature
- Respiratory Rate
- Heart Rate
- Left Finger Pulse Volume
- Right Finger Pulse Volume
- Left Toe Pulse Volume
- Right Toe Pulse Volume
- Left Arm EMG
- Right Arm EMG
- Left Leg EMG
- Right Leg EMG
- Systolic Blood Pressure
- Diastolic Blood Pressure
- Mean Blood Pressure
- Thoracic Impedance
- Stroke Volume
- Cardiac Output
- Total Peripheral Resistance
- Vagal Tone

3.3.1 Heart rate

The heart rate was determined by monitoring the elapsed time between two successive QRS complexes detected in an ECG. The ECG wave form was assigned channel 4 on the WinDaq data acquisition system. The QRS detection algorithm that was used in this system was the one proposed by Pan and Tompkins [13]. The reciprocal of
the time elapsed between successive QRS complexes gave the heart rate in beats per second, and multiplying that by 60 gave the heart rate in beats per minute. The event of QRS detection was critical to the AFTE software system, since it was used as the gating signal for the processing of pulse volume and blood pressure signals.

For the detection of the QRS complex, the basic signal processing structure described in Figure 1.4 was used. The filter coefficients were obtained by designing the required filters in MATLAB 6.5 Filter Design and Application tool. The QRS detection algorithm suggested by Pan and Tompkins [13] has been found to be most robust and efficient as compared to other QRS detection algorithms. The band pass filter to maximize QRS energy was implemented by cascading a low pass and high pass filter together.

**Low-pass filter**

The low pass filter that was used was a sixth order Butterworth filter and its transfer function was

\[
H(z) = \frac{0.0013 + 0.0079z^{-1} + 0.0198z^{-2} + 0.0264z^{-3} + 0.0198z^{-4} + 0.0079z^{-5} + 0.0013z^{-6}}{1 - 2.8384z^{-1} + 3.8236z^{-2} + 2.9383z^{-3} + 1.3398z^{-4} + 0.3393z^{-5} + 0.037z^{-6}}
\]

\[
………… (3.1)
\]

where \( z = e^{j\omega} \)
Figure 3.2 shows the magnitude response of the low-pass filter which has been described in this section. The low-pass filter used was designed such that it gave an attenuation of 40 dB at 60 Hz.

![Magnitude Response graph](image)

Figure 3.2: Magnitude response of low pass filter designed for use in QRS detection algorithm

**High-pass filter**

The high-pass filter that was used in cascade with the low-pass filter to achieve the band-pass characteristics to maximize the QRS energy was a third order Chebyshev I filter. The transfer function of the high pass filter that was used in the monitoring system is
\[ H(z) = \frac{0.8042 - 2.4125z^{-1} + 2.4125z^{-2} - 0.8024z^{-3}}{1 - 2.59z^{-1} + 2.2199z^{-2} + 0.6235z^{-3}} \] (3.2)

where \( z = e^{j\omega} \).

Figure 3.3 shows the magnitude response of the high pass filter.

![Magnitude Response](image)

Figure 3.3: Magnitude response of high pass filter used in QRS detection algorithm

**Derivative filter**

The band pass filtering stage was followed by the derivative stage which acted as a high pass filter. A two point derivative was applied to the band pass filtered signal. This stage amplified the higher frequencies characteristics of the QRS complex while
attenuating the lower frequencies of the P and T waves. The transfer function of the derivative filter that was used is given by

\[ H(z) = 0.1 \cdot (2 + z^{-1} - z^{-3} - 2z^{-4}) \]  

where \( z = e^{j\omega} \)

and the difference equation is given by

\[ y(nT) = 0.1 \cdot (2 \cdot x(nT) + x(nT - T) - x(nT - 3T) - 2 \cdot x(nT - 4T)) \]  

where \( y(nT) \) is the output and \( x(nT) \) is the input to the system.

**Squaring Function**

After differentiation, the signal was squared point by point. The equation of the function used is

\[ y(nT) = [x(nT)]^2 \]  

where \( y(nT) \) is the output and \( x(nT) \) is the input to the system.

This made all data points positive and performed a non linear amplification of the output signal from the derivative emphasizing the higher ECG frequencies.

**Moving Integral**

The purpose of using moving window integration is to obtain wave-form feature information in addition to the slope of the R wave [13]. The moving integral is calculated from

\[ y(nT) = (1/N) \cdot [x((n-(N-1))T) + x((n-(N-2))T) + ....... + x(nT)] \]  

(3.6)
where $y(nT)$ is the output, $x(nT)$ is the input to the system and $N$ is the number of samples in the width of the integration window. According to Pan and Tompkins, the width of the moving window should be the same as the widest possible QRS complex [13]. The value of $N$ was empirically chosen as 32 for a sample rate of 250 samples/s which corresponds to 128 ms.

**Threshold Adjust**

In the Pan-Tompkins algorithm, the threshold is adjusted to float over the noise. The set of thresholds applied were computed from the formulae described by Pan and Tompkins which are described below.

\[
SPKI = 0.125 \, PEAKI + 0.875 \, SPKI
\]

(3.7)

(if $PEAKI$ is the signal peak)

\[
NPKI = 0.125 \, PEAKI + 0.875 \, NPKI
\]

(3.8)

(if $PEAKI$ is the noise peak)

\[
THRESHOLD \, II = NPKI + 0.25 \, (SPKI - NPKI)
\]

(3.9)

where all the variables refer to the integrated waveform.

$PEAKI$ was the overall peak.

$SPKI$ was the running estimate of the signal peak.

$NPKI$ was the running estimate of the noise peak.

$THRESHOLD \, II$ was the threshold applied to the output of the moving average filter.
3.3.2 Respiratory rate

The respiratory waveform was obtained from a strain gauge which senses the expansion and contractions of the chest cavity during breathing. The respiratory waveform was assigned channel 3 on the WinDaq data acquisition system. The respiratory rate from the respiratory waveform was estimated by determining the elapsed time between successive inspiratory cycles. The respiration wave was first subjected to moving average filtering to remove any stray artifacts due to motion and 60 Hz noise. The filtered wave form was then subjected to derivative filtering and rectification. The derivative filter used had the difference equation

\[
y(n) = 2 \cdot x(n) + x(n - 1) - x(n - 3) - 2 \cdot x(n - 4)
\]

(3.10)

where \(y(nT)\) is the output and \(x(nT)\) is the input to the system.

The differentiated signal was rectified and then a moving average filter was applied to it. A threshold was then applied to this rectified signal so that a pulsatile signal was generated. The threshold was determined using the same method as described in Section 2.3.1. The rising edge of this pulsatile signal was treated as the start of the breath. The reciprocal of the time elapsed between successive pulses determines the respiratory rate in breaths per second. The software was designed to display the respiratory rate in breaths per minute.
3.3.3 Finger pulse volumes and toe pulse volumes

Pulsatile signals such as the finger pulse volume and toe pulse volume were processed using the low pass filter described in Section 2.3.1. The left and right finger pulse volumes were assigned channels 1 and 2 respectively on the WinDaq data acquisition system and the left and right toe pulse volume waveforms were assigned channels 6 and 7 on the WinDaq data acquisition system. The purpose of pre-processing the pulse volume signals with a low-pass filter was to eliminate any distortion that may arise due to the 60 Hertz power line interference. The finger and toe pulse volumes were determined by calculating the difference between the base volume, characterized by the valleys in the wave form, and the peak volume characterized by the peaks in the
waveform. Figure 3.5 shows the finger pulse volume as seen on the WinDaq screen with the peak and valley marked.

![Waveform Image]

**Figure 3.5: Blood volume pulse signal**

The arrival of the QRS detection pulse signified an imminent blood volume pulse and was used to trigger the algorithm to search for the peak and valley in the waveform. The peak and valley search algorithms were based on the sample, compare and hold technique where successive points were sampled and compared with the preceding data points. In the case of peak search algorithm, if the value of the data point was found to be greater than the preceding points, it was stored and discarded if the contrary was true. Similarly, in the case of valley search algorithm, if the value of the data point was found to be less than the preceding points, it was stored and discarded if the contrary was true.
3.3.4 Blood pressure

The QRS detection pulse was used as a gating pulse to trigger the peak and valley search algorithm on the blood pressure waveform. The blood pressure wave form was assigned channel 13 on the WinDaq data acquisition system. The peak values of the blood pressure waveform gave the systolic blood pressure and the valleys of the blood pressure waveform represented the diastolic blood pressure. The peak and valley search algorithms that were applied, used the same technique as described in Section 2.3.3. The mean arterial pressure (MAP) was determined by finding the average value of the data points of the blood pressure waveform, between consecutive QRS complexes.

Figure 3.6: Blood pressure signal
3.3.5 Thoracic impedance (Z₀) and dZ/dt

The impedance cardiograph and the dZ/dt signal was used to estimate the stroke volume and cardiac output using Kubicek’s method [1] [8]. A tetra-polar electrode arrangement was used to acquire the impedance cardiograph signal [1]. The stroke volume was determined using Kubicek’s formula which is described below

\[
SV = \rho \cdot \left( \frac{L}{Z_0} \right)^2 \cdot T \cdot \left( \frac{dZ}{dt} \right)_{max}
\]

where,

SV = Stroke Volume

\(\rho\) = the resistivity of blood (Ω-cm)

L = the separation (cm) between the two inner (2, 3) potential measuring electrodes,

\(Z_0\) = the basal impedance between potential measuring electrodes (2, 3),

\((dZ/dt)_{max}\) = the maximum rate of change of impedance (Ω/sec), and

T = Ventricular ejection time

The average resistivity of blood is known to be 159 Ω-cm. The ventricular ejection time was taken to be 0.28 seconds. The distance between the two inner electrodes was obtained from the calibration screen and was usually maintained at 25 cms. The value for \((dZ/dt)_{max}\) was obtained by detecting the positive envelope of the dZ/dt signal.
Figure 3.7 shows the wave forms for dZ/dt and $Z_0$ wave forms. The dZ/dt wave form was filtered using the high pass filter described in Section 2.3.1. This was done to eliminate the base line drift characteristic of the dZ/dt signal. The upper envelope of the high pass filtered dZ/dt signal was then obtained using a sample and hold algorithm. The sample and hold algorithm was reset with every detection of the QRS complex. The peak value of the obtained dZ/dt is used Kubicek’s formula to calculate the stroke volume.

Figure 3.7: dZ/dt and $Z_0$ signal
3.3.6 Total peripheral resistance

The total peripheral resistance (TPR) is the sum total of the impediment to the flow of blood in the blood vessels. In conditions where the blood vessels in the body become constricted, the TPR increases and when the blood vessels in the body dilate, the TPR decreases [22]. In this manner, TPR can serve as an index of patient health and was included as one of the parameters of this monitoring system. The total peripheral resistance was determined from the Mean Blood pressure and the cardiac output using the relationship

\[
\text{Total Peripheral Resistance} = \frac{\text{Mean Blood Pressure}}{\text{Cardiac Output}} [22]
\]

3.3.7 Electromyogram (EMG)

An Electromyogram signal serves as a measure of muscle activity in a person. The purpose of processing of the EMG waveform was to derive an activity index of the muscle. When muscles are active, they produce an electrical voltage that is proportional to the level of muscle activity. For this project, the signal that was acquired is the surface EMG. Surface EMG signals were acquired from left arm, right arm, left leg and right leg of the AFTE trainee/patient. The processing of this signal involved envelope detection, to get an estimate of the amplitude of the Electromyogram. The envelope detection was performed by using a rectification [32] function described by

\[
y(nT) = |x(nT)|
\]  \hspace{1cm} (3.12)

where \(y(nT)\) is the output and \(x(nT)\) is the input to the system.
A moving average window was applied to the output of the rectification function. The function of moving average is described by

\[ y(nT) = \frac{1}{N} \cdot [x(nT - (N - 1)T) + x(nT - (N - 2)T) + \ldots + x(nT)] \]  

(3.13)

where \( y(nT) \) is the output and \( x(nT) \) is the input to the system. For EMG processing, the value of \( N \) was empirically determined to be 50 which corresponds to 200 ms at a sample rate of 250 samples/s.

3.3.8 Skin temperature

The waveform that represents the skin temperature is a slow moving waveform. The processing of the temperature signal involved preprocessing with a low-pass filter followed by a moving average filter having a width of 1000 samples. The moving average filter was used to eliminate any high frequency components that may have been introduced in the signal due to motion or any other source of noise.

3.3.9 Skin conductance level

Skin conductance level signal, as in the case of the temperature waveform, is a relatively slow moving signal. A moving average filter of 125 data points was applied to the Skin conductance signal, which corresponds to 500 ms at a sampling rate of 250 samples per second.
3.3.10 Vagal tone

One of the parameters that the physiological training and monitoring software estimated was vagal tone. The frequency analysis of heart rate shows two spectral peaks, at 0.04-0.15 Hz and 0.15-0.4 Hz [30]. This software measured the high frequency bin of heart rate variability, which is referred to as vagal tone.

The first step in processing heart rate signal for vagal tone was to decimate it to 5 Hz. The decimated heart rate signal was then windowed using a hanning window function and the Fourier transform was determined. The magnitudes in the frequency bin of 0.15 HZ and 0.45 Hz were determined and an average of the values was determined. This gave the average power spectral density at these frequencies and this was reported as the vagal tone.

For vagal tone processing, a frequency resolution of at least 0.01 Hz was required. Therefore, 100 seconds worth of data was required for efficient estimation of vagal tone. Since the heart rate signal had been decimated to 5 Hz, 500 sample points were required. To these points a hanning window function was applied to prevent harmonics. For implementation of Fast Fourier Transform (FFT) algorithm [23], it was required that the number of sample points be a power of 2. In this way, a 512 point FFT was implemented which corresponded to 102.4 seconds giving a frequency resolution of 0.01 Hertz.
3.4 Communication system

An efficient communication system was needed for coordinated interaction between the server and client, for the physiological monitoring and training software to work effectively. The TCP/IP was chosen as the preferred communication protocol over the Internet for the interaction between the client and server of the system, due to its robustness and the capability of guaranteed data transmission. The Microsoft Winsock ActiveX Control, available in the Microsoft Visual Basic 6.0® components, was used for the purpose of socket programming. The terminal where the data was acquired from the patient and processed, acted as the server. The client end was not designed to perform any signal processing but provided tools to control the server and view the processed data from a distant location. Twenty one ports, port 5837 through port 5857, were used for communication purposes. The five remaining ports were used to transmit the processed data, the status of the displays on the server and the client end, the auditory feedback signals and the synchronization signals.

3.4.1 Transmission of raw data

Sixteen ports, 5840 through 5855, were used to transmit the unprocessed raw data from the server end to the client end. At a sampling rate of 250 samples/second/channel, there is one data point to be transmitted for each channel every 64 ms. For such high speed data de-interlacing and transmission, Mabry® HITIME, a high resolution timer, was used. The high resolution timer triggered every 64 ms and one data point was sent to the client from each port.
3.4.2 Transmission of processed data

Port 5856 was assigned for the transmission of processed data. The 20 values of processed data were transmitted in the form of a byte array over the Internet every 256 ms. A 256 ms timer was used to sample the processed data and transmit it to the client.

3.4.3 Transmission of information to update the displays

Port 5857 was used for transmission of information regarding change in the screen display status. For example, when a panel is clicked to highlight a display on the subject screen, either on the server system or the client system, information is sent on port 5857 to update the screen on either end. This information is also transmitted in the form of a byte array.

3.4.4 Transmission of signals to control WinDaq® data acquisition software

The software was developed so that the WinDaq data acquisition software could be controlled using the physiological monitoring and training software, both from the server end and the client end. Port 5837 was dedicated to transmit the information for controlling the WinDaq data acquisition software.

3.4.5 Transmission of synchronization information

Port 5838 was dedicated to exchange information to synchronize the client and the server software. This was required so that both the ends have the same status on displays and synchronization variable.
3.4.6 Transmission of information on detection of QRS complex

The detection of a QRS complex corresponds to the occurrence of a heart beat. Port 5839 was dedicated to transmit information when a QRS complex was detected. This information was used to make an LED like display blink with every QRS complex that was detected, which provided a visual indication of the heart rate.

3.5 Graphical user interface and other features

A graphical user interface (GUI) was needed to make the software user friendly and features were added to the software for the purpose of feedback training and warning systems in cases of extreme conditions in patients. On the server end, where the patient or the trainee was present, three monitors were slaved to one computer system. One of the monitors was for display of processed data to the investigator present at the local site and another monitor was for the local investigator to view the raw data using the WinDaq data acquisition software display. A third screen was for the display of processed data output to the patient or the trainee. The client software was also developed to have an investigator screen and a screen to view the raw data that being acquired, similar to the ones at the server end. The ActiveX controls from the IOCOMP® plot pack and IOCOMP® professional pack were used to develop the GUI for the software.

The investigator screen was designed to have 20 panels each displaying one of the 20 indices of health that are determined using the algorithms described in Section 3.3. Figure 3.8 shows the investigator screen which displays the processed data.
3.5.1 Investigator toolbar

The first button on the toolbar with the computer network icon, when clicked, brought up the connection window as shown in Figure 3.9. On clicking on the “Allow Client to Connect” button, after entering the appropriate port number, 21 ports were opened and were made ready to accept connection requests from the clients. The connection status is displayed on the investigator screen as “Listening”, “Connected”, or “Not Connected”.

Figure 3.8: Investigator screen
Figure 3.9: The connection window

Clicking of the button on the tool bar with the oscilloscope symbol on it, brought up a window which could be used to control the raw signal display on the WinDaq data acquisition software. This is illustrated in Figure 3.10.
Figure 3.10: Control of WinDaq data acquisition and display – WinDaQ properties window

The up and down buttons in the compression control panel of the WinDaq Properties window, were used to increase or decrease the compression on the WinDaq screen. The Scroll up and Scroll down buttons in the WinDaq channel control panel were used to scroll through the channels on the Windaq screen. It was also possible to determine the number of channels visible at a time on the WinDaq screen by entering the desired number in the text box and clicking on the Number of Channels button.

The folder button on the tool bar was used to set up a file for recording a data file. Two files were set up for recording, WinDaq file containing the raw waveform with the *.wdq extension and a text file containing the processed data recorded every time a QRS
complex was detected. The buttons with the floppy diskette and the crossed out floppy diskette were used to start and stop recording of the files respectively. The recording status is displayed on the investigator screen as “Standby”, “Recording” and “Not Recording”. The file name with path is also displayed on the investigator screen.

The button with the check mark on it, on the toolbar was used to bring up the calibration screen. This screen also showed every time the physiological monitoring software was started up. This screen was used to set up the scaling factors for the 20 physiological parameters that were being measured. The Figure 3.11 shows the calibration screen.
Figure 3.11: Calibration screen

The button, on the toolbar, with the ear was used to control the auditory feedback tools that could be used in feedback training or as an alarm system in the physiological monitoring system. The screen that is displayed on clicking tone control button is shown in Figure 3.12. The button with the stop sign was used to exit from the physiological monitoring software and terminate the program.
3.5.2 Display control of the patient screen

The trends of the various output parameters can be viewed graphically by clicking on the processed data panels, dragging and dropping it on the empty panels on the right side of the investigator screen. The trends displays are shown in Figure 3.13.
Figure 3.13: Trends of the processed data outputs – shown on the right of the investigator screen

The displays on the patient screen were allowed to be controlled by the investigator. Any processed data display could be made available to the patient or feedback trainee by clicking on the panel. The background of the panel which was displayed to the patient or trainee was made to turn dark, to indicate the same. This is illustrated in Figures 3.14 and 3.15.
Figure 3.14: Panel display of parameters selected to be displayed to the patient, as seen on the investigator screen
Figure 3.15: Panel display of parameters selected to be displayed to the patient, as seen on the patient/subject screen.

The vertical bar displays could be made visible to the patient or trainee by clicking on the LED button on each of the panel. When the vertical bar was displayed to the patient/trainee, the LED button glowed to indicate the same. This is illustrated in Figures 3.16 and 3.17.
Figure 3.16: LED button shown glowing when vertical bar selected to be displayed for heart rate on the patient screen.
3.5.3 Tones for feedback training

The tones for feedback training are sounded in 3 modes.

(i) Plays sound when the readings are above or below the set limits and sound turns off when the readings are within the set limits.

(ii) Plays sound when the readings are within the set limits and sound turns off when readings go above or below the set limits.

(iii) The frequency of the sound being played varies with any change in the readings.
The desired mode could be selected by clicking on the radio buttons labeled as these modes on the Tone Control screen shown in Figure 3.12. The frequency of the tone sounded could also be set on the Tone Control screen. The sounding of the tones of different frequencies was accomplished using the Raaga® ActiveX control. The upper and lower limits within which the tones work can be set by adjusting the white and the yellow pointer as shown in Figure 3.18. The upper and lower limit are indicated on the patient screen by showing different colors for the region within the limit than that which is outside the limit. This is illustrated in Figure 3.20. Fine adjustment of the set limits was possible double-clicking on the vertical bar and bringing up the screen shown in Figure 3.19. Patients or subjects can be trained for two parameters at a time by sounding tones. The tones are turned on by clicking on the button with the crossed out speaker at the bottom left hand corner of each display panel as shown in Figure 3.21. It was made possible to switch on a maximum of two tones at a time.
Figure 3.18: Setting of limits using the yellow and white arrows on the heart rate panel, for the sounding of tones
Figure 3.19: Fine adjustment to set tone limits
Figure 3.20: Set limits for sounding tones, visible as different colored regions on patient screen
3.5.4 Voice commands for feedback training

Voice commands were also required for feedback training exercises involving tilting and swaying the head according to instructions. These were provided by storing audio files on the computer memory and playing them using the functions provided in the windows API library[19][23]. These voice instructions could be provided either in English or in Russian. The voice commands were turned on by clicking on the check box indicated in Figure 3.22. The radio buttons titled English and Russian were used to select the language for the voice commands.
3.5.5 Metronome

It was possible to turn on a periodic beep, termed as the metronome, to help the trainee maintain a steady respiration rate. The metronome was turned on by clicking on the check box labeled as Metronome, shown in Figure 3.23, on the investigator screen. It was made possible to control the rate at which the beep was sounded by adjusting the metronome frequency on the tone control screen shown in Figure 3.12.
3.6 Statistical analysis

Paired t-tests were performed on each of the parameters that were estimated using the physiological monitoring software system and parameters that were determined manually, to determine the presence of any statistically significant differences between the actual value of the parameters and the values estimated by the software. The SAS® software was used for performing the statistical analysis. A value of $\alpha = 0.05$ was chosen for performing the paired t-tests.
CHAPTER IV

RESULTS

The results from the techniques prescribed in the previous section are presented in this chapter. These include the intermediate output during the real-time execution of the implemented algorithms.

4.1 Performance analysis at intermediate stages

This section presents the output of the intermediate stages of the signal processing algorithms that were implemented for real-time processing. The intermediate results were used to verify performance of the algorithms at each stage of implementation.

4.1.1 Heart rate

The output, at each stage of the Pan-Tompkins QRS detection algorithm that was implemented in the physiological monitoring software, is shown in Figure 4.1. It can be observed from the figure that by passing the ECG signal through a band pass filter, the P and the T waves were suppressed. The QRS wave was then enhanced by passing the ECG wave through the derivative filter and a squaring function. A threshold function was then applied to the output of the moving average filter which resulted in a pulse train like
signal where each pulse signified the detection of a QRS complex. The elapsed time between two consecutive pulses was measured and the heart rate was determined.

Figure 4.1: Stages in the implementation of the Pan-Tompkins algorithm
Figure 4.2: Plot of the manually determined heart rate and heart rate measured by monitoring software.

4.1.2 Respiratory rate

Figure 4.4 depicts intermediate outputs for the respiration signal processing algorithm. The figure shows the original respiration signal, the low pass filtered signal with stray noise removed, the half wave rectified derivative of the respiratory signal and the smoothed derivative signal. When the smoothed derivative function was subjected to thresholding, it gave rise to a pulsatile signal as shown in Figure 4.4, which was used to estimate the respiratory rate.
Figure 4.3: Stages in the implementation of breath detection algorithm

Figure 4.4: Plot of the manually determined respiratory rate and respiratory rate measured by monitoring software.
4.1.3 Blood volume pulse

The output at each stage of the implementation of the algorithm to determine blood volume pulse is shown in Figure 4.5. The low pass filter that was implemented removed the stray noise in the signal. The positive and negative envelopes of the blood volume pulse were determined and the difference between the upper and lower envelopes of the blood volume pulse gave the pulse volumes. Figure 4.5 shows the real time plots of the original pulse volume signal, the envelopes of the pulse volume signal and the plot of the pulse volume signal, which were used to test the implementation of the algorithm. A graphical comparison of the blood pulse volumes determined by manual inspection and that determined by the software is illustrated in Figure 4.6.
Figure 4.5: Stages in the determination of pulse volume

Figure 4.6: Plot of the manually determined blood volume pulse and blood volume pulse measured by monitoring software.
4.1.4 Blood pressure

Figure 4.7 shows the results of the implementation of the algorithm to process the blood pressure signal. The figure shows the plot of the upper envelope which represents the systolic blood pressure, the lower envelope which represents the diastolic blood pressure and the mean blood pressure or the mean arterial pressure. These plots were used to evaluate the performance of the software in real time. Graphical comparisons of the systolic blood pressure, diastolic blood pressure and mean blood pressure measured using the software and the values of the same obtained by manual inspection are shown in Figures 4.8, 4.9 and 4.10 respectively.

![Blood Pressure Processing](image)

**Figure 4.7:** Stages in the determination of systolic, diastolic and mean blood pressure
Figure 4.8: Plot of the manually determined systolic blood pressure and systolic blood pressure measured by monitoring software.

Figure 4.9: Plot of the manually determined diastolic blood pressure and diastolic blood pressure measured by monitoring software.
Figure 4.10: Plot of the manually determined mean arterial pressure (MAP) and MAP measured by monitoring software.

4.1.5 Impedance cardiogram– dZ/dt and $Z_0$

Figure 4.11 shows the plots of the raw dZ/dt signal, the high pass filtered dZ/dt signal and the envelope of the dZ/dt signal. The high pass filter is applied to remove the base line drift associated with the dZ/dt signal. These plots were used to evaluate the performance of the software in real time. The graphical comparisons, of $Z_0$ measured using the software and the values of the same obtained by manual inspection, is shown in Figures 4.11.
Figure 4.11: Stages in the determination of envelope of $dZ/dt$ envelope

Figure 4.12: Plot of the manually determined $Z_0$ and $Z_0$ measured by monitoring software.
4.1.6 Stroke volume and cardiac output

The stroke volume is measured using Kubicek’s formula. The values of $Z_0$ and envelope of $dZ/dt$ are obtained using techniques described in Section 2.3.5. The stroke volumes and the cardiac outputs obtained by manually examining the $dZ/dt$ and $Z_0$ waveforms and the values obtained using the software are plotted and compared in Figures 4.13 and 4.14 respectively.

![Figure 4.13: Plot of the manually determined stroke volume and stroke volume measured by the monitoring software.](image-url)
4.1.7 Total peripheral resistance

The total peripheral resistances are obtained by manually calculating cardiac output and mean blood pressure and also from the software. The two sets of values are plotted and compared in Figure 4.15.
Figure 4.15: Comparison of total peripheral resistance determined by the software and that determined by manual inspection

4.1.8 Electromyogram

The electromyogram graphs obtained by manually inspecting the EMG waveform and the values obtained from the software are plotted. The plot is shown in Figure 4.16 and compares the two sets of values.
Figure 4.16: Comparison of left arm electromyogram determined by the software and left arm electromyogram determined by manual inspection

4.1.9 Skin conductance level and temperature signal

Figures 4.17 and 4.18 show the comparison between sets of values of skin conductance level and skin temperature. One set of data is obtained using the monitoring software and the other set of values is obtained by manual examination of raw data.
Figure 4.17: Comparison of skin conductance level

Figure 4.18: Comparison of temperature
4.2 Statistical analysis

Paired t-tests ($\alpha=0.05$) were performed on the data obtained as the output of the various algorithms that were implemented in the software and the manually calculated values of the same parameters. Data were recorded from the software for duration of one minute and the analysis was performed on these data. The t-values and the probabilities for the measured parameters are listed in Table 4.1. Based on the calculation shown in Table 4.1, it was concluded that the parameter values measured by the physiological monitoring software are not significantly different from the parameter values measured manually.
Table 4.1: Paired t-Test for the 20 channels of output

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Number of parameters (N)</th>
<th>t – value</th>
<th>$P_r$</th>
<th>Significant difference at $\alpha=0.05$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Heart rate</td>
<td>80</td>
<td>1.0240</td>
<td>0.3071</td>
<td>No</td>
</tr>
<tr>
<td>Respiratory rate</td>
<td>78</td>
<td>-0.3383</td>
<td>0.7356</td>
<td>No</td>
</tr>
<tr>
<td>Pulse volume</td>
<td>80</td>
<td>0.3086</td>
<td>0.7579</td>
<td>No</td>
</tr>
<tr>
<td>Systolic blood pressure</td>
<td>80</td>
<td>-0.4130</td>
<td>0.6802</td>
<td>No</td>
</tr>
<tr>
<td>Diastolic blood pressure</td>
<td>80</td>
<td>0.1535</td>
<td>0.8782</td>
<td>No</td>
</tr>
<tr>
<td>Mean blood pressure</td>
<td>69</td>
<td>-0.0220</td>
<td>0.9373</td>
<td>No</td>
</tr>
<tr>
<td>Thoracic impedance ($Z_0$)</td>
<td>75</td>
<td>0.4702</td>
<td>0.6389</td>
<td>No</td>
</tr>
<tr>
<td>Stroke volume</td>
<td>75</td>
<td>-0.4373</td>
<td>0.6625</td>
<td>No</td>
</tr>
<tr>
<td>Cardiac output</td>
<td>74</td>
<td>-0.6093</td>
<td>0.5433</td>
<td>No</td>
</tr>
<tr>
<td>Total peripheral resistance</td>
<td>69</td>
<td>-1.0126</td>
<td>0.3130</td>
<td>No</td>
</tr>
<tr>
<td>Electromyogram</td>
<td>80</td>
<td>1.061</td>
<td>0.2903</td>
<td>No</td>
</tr>
<tr>
<td>Skin conductance level</td>
<td>80</td>
<td>0.1617</td>
<td>0.8718</td>
<td>No</td>
</tr>
<tr>
<td>Temperature</td>
<td>80</td>
<td>-0.0340</td>
<td>0.9730</td>
<td>No</td>
</tr>
</tbody>
</table>
CHAPTER V
DISCUSSION AND CONCLUSIONS

5.1 Errors in data acquisition

As mentioned in Section 3.2, the DI-720 series of instruments have a 16 bit A/D converter for higher resolution measurement accuracy but the maximum accuracy that can be obtained using a WinDaq software is 14 bits. The two least significant bits are truncated to make the WinDaq software computationally efficient. This leads to truncation error in the values of the acquired data.

A quantization error is always associated with any analog to digital converter. Quantization error is the error that is caused due to the resolution of the A/D converter. The resolution of the A/D converter can be improved by increasing the number of bits. The higher the resolution, the lower is the value of quantization error. The amplitude resolution of the A/D converter is given by $\frac{1}{2^{n-1}}$, n is the number of bits in the A/D converter. In the case of WinDaq, a dynamic range of +10 to -10 Volts at a gain of 1 was used and 16 bits A/D converter was used. Therefore the dynamic range was calculated as

$$20 \cdot \frac{1}{2^{15}} \approx 0.61mV$$
5.2 Errors introduced during signal processing

There is error introduced due to quantization noise when digital filters are used. Quantization noise is the error introduced due to finite precision of registers or variables storing the intermediate values obtained during the implementation of the digital filters.

“Two consequences of such an error are (1) The state variables that make up the filter can only represent an integral multiple of the smallest quantum, and (2) There is a maximum value that the register can represent in a one-to-one correspondence. The first effect is known as quantization, the second, as overflow (Oppenheim and Schafer, 1975; Roberts and Mulis, 1987).” [28]

5.3 Conclusion

Based on the results discussed in the Chapter IV, it was concluded that a 16 channel input, 20 channel output remotely controlled physiological monitoring and feedback training system was successfully implemented. The algorithms that were implemented in the monitoring system software, efficiently and successfully determined the 20 indices of health that were to be calculated. A delay of about 1 to 2 seconds was observed between the actual occurrence of the event and the recording of the event. This delay was expected as the algorithms would use some processing time to perform the various calculations and comparisons. A remote client was also designed that could be used to control the displays, data recording and acquisition, the various feedback training tools on the server side of the monitoring system.
Based on the above study and statistical analysis, we fail to reject the null hypothesis and conclude that the interactive remote physiological monitoring and feedback training system can be used to faithfully monitor changes in the 20 physiological parameters listed and train subjects for autogenic feedback training, i.e. $\mu_1 = \mu_2$.

5.4 Limitations of the current study

In this study, a single server single client system was developed. When the TCP/IP protocol is used, the order arrival of data points at the client end need not necessarily be the same as the order in which the data points were sent. Due to this inherent drawback of the TCP/IP protocol, when it was attempted to send the raw data points through a single port, the integrity of the individual channels was lost. Consequently, to maintain the integrity of the channels, 16 different ports were used to transmit the raw physiological data. This loaded the processor and led to delays in the processing of physiological signals. A method needs to be developed to reduce the number of ports used for transmitting the raw data.

For the same reason that has been stated above, the remote monitoring system was limited to a one server – one client system. If the number of ports was reduced, a system involving single server and multiple clients can be developed.

The respiratory signal processing was based on simple filtering and averaging techniques. The respiratory rate reported may sometimes be erroneous as the extent of inspiration or expiration varies from person to person. A simple filtering algorithm for
such a wave form may not be the ideal one. Adaptive filtering techniques or neural network based methods to detect and differentiate between artifacts and genuine breath signals may be better suited for respiratory signal processing.

5.5 Future work

As mentioned in Section 5.4, the software can be improved in such a way that fewer channels are used for the data communication between server and client. This would enable a single server – multiple client system to be implemented. One of the 16 channels of that were acquired was left unused. This channel can be used to acquire one more raw physiological signal and processing of this signal can be added to the current software. This is possible since the software has been implemented in a modular fashion and addition of another module is not difficult. With the improvement in computer technology and reduced size of computer hardware, the monitoring system can be implanted without much modification to work on devices such as Palmtops or Personal Digital Assistants (PDA’s).
REFERENCES


APPENDICES
APPENDIX A

PROGRAM FOR IMPLEMENTATION OF THE PATIENT/SUBJECT SIMULATOR

#include <stdio.h>
#include <conio.h>
#include <dos.h>
#include <iostream.h>

#define BASE1 0x300
#define BASE2 0x320

#define CH1 BASE1+0
#define CH2 BASE1+2
#define CH3 BASE1+4
#define CH4 BASE1+6
#define CH5 BASE1+8
#define CH6 BASE1+10
#define CH7 BASE1+12
#define CH8 BASE1+14

#define CH9 BASE2+0
#define CH10 BASE2+2
#define CH11 BASE2+4
#define CH12 BASE2+6
#define CH13 BASE2+8
#define CH14 BASE2+10
#define CH15 BASE2+12
#define CH16 BASE2+14

void main(void)
{
    //******SETTING CARDS TO AUTOMATIC UPDATE MODE***************
    //******AND FULL SCALE VOLTAGE RANGE (see manual)**************
    inp(BASE1+10);
    inp(BASE1+15);
}
inp(BASE2+10);
inp(BASE2+15);

JECTION PROPERTIES
char *fname;
int n[16];
float v;
FILE *fp;
cout<<n'nn******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************n******************************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outport(CH12,n[11]);
outport(CH13,n[12]);
outport(CH14,n[13]);
outport(CH15,n[14]);
outport(CH16,n[15]);

delay(7);

}

fclose(fp);
cout<<"nBye";
APPENDIX B

HARDWARE USED IN DATA ACQUISITION FOR AFTE

<table>
<thead>
<tr>
<th>Signal</th>
<th>Body Location</th>
<th>Equipment</th>
</tr>
</thead>
<tbody>
<tr>
<td>Blood volume pulse 1</td>
<td>left index finger</td>
<td>NASA</td>
</tr>
<tr>
<td>Blood volume pulse 2</td>
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</tr>
<tr>
<td>Blood volume pulse 3</td>
<td>not used</td>
<td></td>
</tr>
<tr>
<td>Blood volume pulse 4</td>
<td>left little finger</td>
<td>AFS-2</td>
</tr>
<tr>
<td>respiration</td>
<td>chest</td>
<td></td>
</tr>
<tr>
<td>Electrocardiography</td>
<td>chest - 3 electrodes</td>
<td></td>
</tr>
<tr>
<td>skin temperature</td>
<td>left little finger</td>
<td></td>
</tr>
<tr>
<td>skin conductance</td>
<td>left palm - 2 electrodes</td>
<td></td>
</tr>
<tr>
<td>Electromyography 1</td>
<td>left forearm - 3 electrodes</td>
<td>J &amp; J Electronics</td>
</tr>
<tr>
<td>Electromyography 2</td>
<td>right forearm - 3 electrodes</td>
<td></td>
</tr>
<tr>
<td>Electromyography 3</td>
<td>left calf - 3 electrodes</td>
<td></td>
</tr>
<tr>
<td>Electromyography 4</td>
<td>right calf - 3 electrodes</td>
<td></td>
</tr>
<tr>
<td>systolic blood pressure</td>
<td>right wrist</td>
<td>Colin Medical</td>
</tr>
<tr>
<td>diastolic blood pressure</td>
<td></td>
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</tr>
<tr>
<td>Impedance cardiography</td>
<td>neck and thorax</td>
<td>Bioimpedance Inc.</td>
</tr>
<tr>
<td>Impedance cardiography</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
APPENDIX C

PROOF OF IRB APPROVAL OBTAINED BY NASA Ames Research Center

262-2

Bruce Taylor, Ph.D.
3508 Charming Cross Dr.
Stow, OH 44224

March 23, 2006

Dear Dr. Taylor:

Enclosed is a copy of the e-mail messages sent regarding NASA IRB approvals for your use of data collected as part of a NASA grant. Dr. Ralph Pelligrino, Chairman of the NASA-Ames Human Research Institutional Review board concurred that these data may be used as long as the subjects’ identities are not revealed.

Hope this meets your board’s approval.

Sincerely,

Patricia S. Cowings, Ph.D.
Research Scientist
Date: Tue, 14 Mar 2006 08:35:34 -0800
To: Ralph Pelligra <Ralph.Pelligra-1@nasa.gov>
From: pcowings <Patricia.S.Cowings@nasa.gov>
Subject: Re: IRB Permissions
Cc: Bruce Taylor <bruce@ghoticreek.com>

Thanks Ralph. I am sending a copy of your concurrence below to Bruce Taylor for his records.

Dr. Pat

At 08:27 PM 3/13/2006, you wrote:

Concur- providing the subjects cannot be identified, either directly or indirectly.... R.Pelligra

Bruce

I believe that no additional IRB approvals are necessary if your students are using previously collected data and the files do not reveal the identity of the subject. Since you have
2 files already for subject O27, I've put the raw and processed files of the same subject that were collected during an Autonomic Function Test 027ANS on EOS.

These data were collected under The Morehouse Protocol HRI-242 "Telemedicine Applications of Autogenic-Feedback Training Exercise as a Treatment for Specific Patient Populations," which is scheduled to expire April 7, 2006. We will be closing out that protocol as the grant period has expired.