



Intelligent Sensing Anywhere

# Look4MySounds

**A Remote Monitoring  
Platform for Auscultation**

*Integrated Master in  
Biomedical Engineering*

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Technology*

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**FCTUC**

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# Abstract

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Cardiovascular diseases are the 21st century epidemic. Ageing, obesity, sedentary lifestyle and numerous other factors contribute to its growing numbers, with devastating causes, both economic and social.

Cardiac auscultation is one of the oldest methods for heart function assessment. It is a non invasive, low cost method which provides accurate information about heart mechanics and hemodynamics.

In recent years, remote monitoring technologies have emerged and proven to be very efficient, reducing the burden of healthcare units, reducing hospitalizations, providing better healthcare service and improving patients quality of life. Moreover, cardiac patients are usually high-risk patients, needing constant observation. The use of a cardiac remote monitoring device would reduce travels to healthcare units, by constantly monitoring the patient and detecting a potential problematic event before it would evolve to a critical condition.

The objective of this project is to develop a portable platform for cardiac auscultation and automatic detection of pathologies. As an initial approach, an algorithm for detection of S1 and S2 heart sounds was developed and tested with two sets of data corresponding to pathologic and non-pathologic heart sounds. The obtained results consisted of an average Sensitivity and Specificity of 92.8% and 92.1% for S1 heart sounds and 96.8% and 93.9% for S2, respectively.

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# Resumo

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As doenças cardiovasculares são a epidemia do Século XXI. O envelhecimento populacional, a obesidade, os estilos de vida sedentários e diversos outros factores contribuem para o seu crescimento, com um impacto económico e social devastador.

A auscultação cardíaca é um dos mais antigos métodos para avaliação da função cardíaca. É um método barato e não invasivo, que permite observar com precisão fenómenos cardíacos mecânicos e hemodinâmicos.

Recentemente, as tecnologias de monitorização remota de sinais vitais têm provado ser bastante eficientes quer a nível económico quer a nível de prestação de cuidados de saúde, reduzindo hospitalizações, evitando deslocamento de pacientes a unidades hospitalares, e ao mesmo tempo melhorando a sua qualidade de vida. No caso de doentes cardíacos, é fulcral uma observação constante, já que se tratam de pacientes de risco elevado, pelo que o uso de tecnologias de monitorização remota permitiria uma observação regular e permanente, detectando potenciais eventos problemáticos antes de se tornarem casos críticos.

Este projecto consiste no desenvolvimento de uma plataforma remota para auscultação cardíaca e detecção automática de patologias. Como abordagem inicial a este problema foi desenvolvido um algoritmo para a detecção dos sons cardíacos S1 e S2. Os resultados obtidos foram Sensibilidade e Especificidade médias de 92.8% e 92.1% para sons S1 e 96.8% e 93.9% para S2.

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# Acronyms

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<b>AP</b>	Atrial Pressure
<b>AV</b>	Atrioventricular Valves
<b>CVDs</b>	Cardiovascular Diseases
<b>CWT</b>	Continuous Wavelet Transform
<b>DWT</b>	Discrete Wavelet Transform
<b>ECG</b>	Electrocardiogram
<b>GPRS</b>	General Packet Radio Service
<b>GUI</b>	Graphical User Interface
<b>IDWT</b>	Inverse Discrete Wavelet Transform
<b>MMC</b>	MultiMedia Card
<b>PCG</b>	Phonocardiogram
<b>SD</b>	Secure Digital
<b>SE</b>	Sensitivity
<b>SP</b>	Specificity
<b>VP</b>	Ventricular Pressure

# 1. Introduction

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## 1.1. Motivation

Cardiovascular diseases (CVDs) are the leading cause of death worldwide. Obesity, population ageing, sedentary lifestyles, the rise of chronic diseases incidence and many other factors continue to increase this epidemic, with devastating causes. The cost of CVDs in Europe is estimated to be almost €200 billion each year, and is expected to rise. [1]

The use of remote monitoring devices in cardiology has been proved to significantly improve efficiency of healthcare services, not only reducing overall costs but also improving patients' quality of life. [2] Most devices for cardiac remote monitoring are based on ECG, however, as opposed to cardiac auscultation, it doesn't provide information on heart valves or hemodynamics, important elements for heart diseases detection, especially valvular disorders. Additionally, cardiac auscultation is non-invasive, low cost, reliable and easy to perform.

The possible detection of a problem before it develops into a critical condition has huge benefits for the patient, and could be achieved with just a simple, regular examination done at home. Furthermore, all clinical data can be easily stored and accessed, while the automatic detection of pathologies could assist clinicians in diagnosis.

## 1.2. Objectives

The objective of this project is to develop a low cost, portable platform for auscultation and automatic detection of pathologies. The platform has an integrated stethoscope with which the auscultation sounds are recorded. The platform has local processing capabilities and should be able to automatically detect pathologies. A sample of the recorded sound as well as the diagnosis are stored in a local memory card and remotely sent to a clinician.

## 1.3. Document structure

This document is organized in 9 chapters. Chapter 1 gives a succinct overview of the project, the motivation and objectives.

Chapter 2 presents the team members involved in the project, planned tasks and its temporal evolution.

Chapter 3 presents the theoretical background involved in the project, such as heart anatomy, origin and characteristics of heart sounds and cardiac auscultation.

Chapter 4 analyzes the background problem, namely material and human costs, and presents a few devices for remote monitoring already in the market.

In chapter 5, the Look4MySounds prototype platform specifications and architecture are described.

Chapter 6 describes tools for signal processing such as Fourier and Wavelets Transforms, and techniques commonly used in heart sounds processing.

Chapters 7 and 8 describe the development of the algorithm and the obtained results, respectively.

Finally, chapter 9 presents the conclusions on the developed work as well as possible future improvements to the presented algorithm.

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## 2. Project Management

The project team consisted of a Biomedical Engineering student and his supervisors. The project was developed under a partnership between CEI – Instrumentation Center from the Physics Department of the University of Coimbra, whose research areas include Nuclear and Atomic Instrumentation, Biomedical Instrumentation, Optical Signal Processing, Instrumentation for Plasma Physics and Telemetry and Industrial Control – and ISA – Intelligent Sensing Anywhere, an international company created in 1990 as a spin-off company of the University of Coimbra, specialized in Telemetry and Machine to Machine Communications. In 2008, ISA started developing solutions for the Health market.

The project was developed at CEI and ISA installations, where occasional meetings with Engineer Soraia Rocha and Engineer Catarina Pereira took place.

### 2.1. Project members

The next table shows the members involved in the project and the respective role.

Name	Role
Tiago Sapata	Student
Engineer Soraia Rocha	Project Manager
Engineer Catarina Pereira	Supervisor
Professor Carlos Correia	Project Coordinator
Professor José Basílio Simões	Project Coordinator

**Table 1** Members involved in the project and their respective role.

## 2.2. Tasks

The first trimester was essentially for knowledge acquisition where the student researched heart sounds, algorithms for heart sounds processing, Matlab tools for signal processing and C programming language.

The initial planning was to develop an automatic method for detection of cardiac pathologies in Matlab and convert the code to C programming language to implement it on a microcontroller, however, with the growing complexity of the problem, the initial planning was altered in the mid of the second trimester. The new planning consisted on developing an algorithm for automatic detection the first and second heart sounds in Matlab and the implementation on a microcontroller was cancelled.

In the second and third trimesters the student developed the planned algorithm and collected non-pathologic and pathologic heart sounds at Cardiology Department of Hospital dos Covões for testing and tuning of the algorithm.

## 2.3. Planning

The next image shows the tasks carried out by the student and the respective duration.

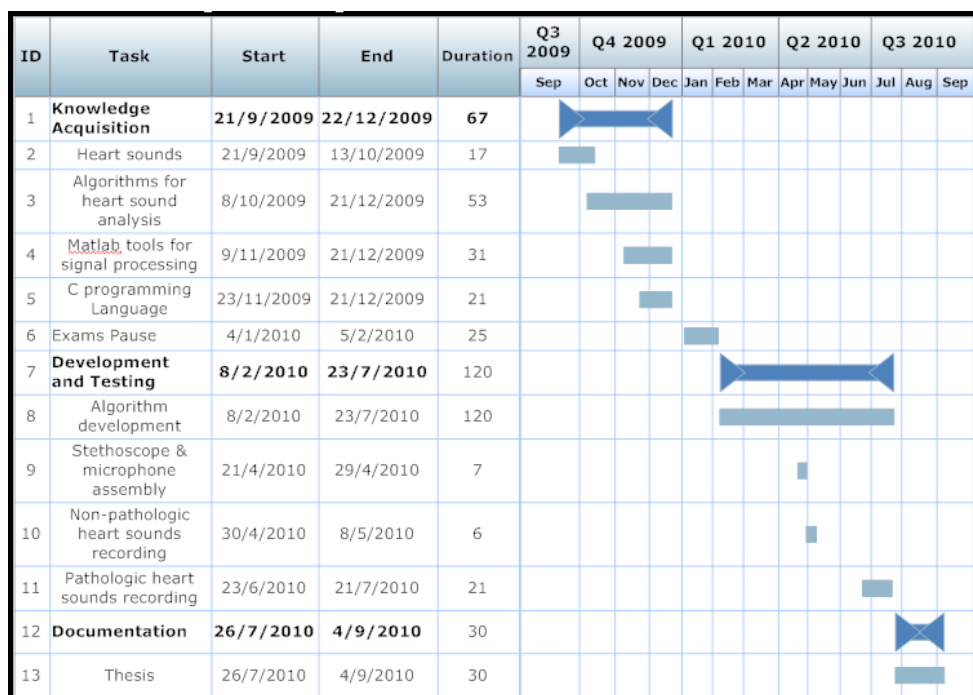


Figure 1 Project's Gantt diagram

## 3. Theoretical Background

### 3.1. Anatomy of the Heart

The heart is one of the most important organs in the human body, providing a constant blood flow which supplies tissues with nutrients and oxygen, while removing unwanted substances such as carbon dioxide and other by-products.

Its structure consists of four chambers: one atrium and one ventricle on each side, called the left heart and the right heart. The atria and ventricles are separated by atrioventricular (AV) valves, tricuspid valve for the right heart and mitral valve for the left heart. Semilunar valves separate the right ventricle from the pulmonary artery and the left ventricle from the aorta, and are called pulmonary and aortic valves, respectively. Heart valves play a major role in the cardiac cycle, preventing blood from back flowing.

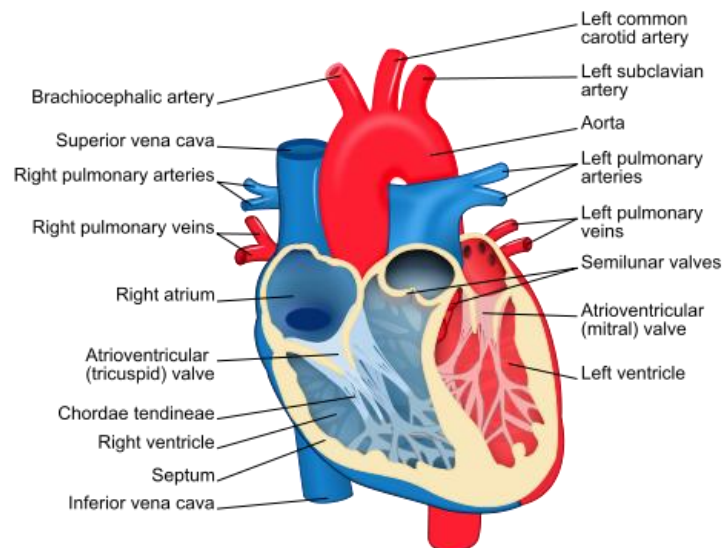
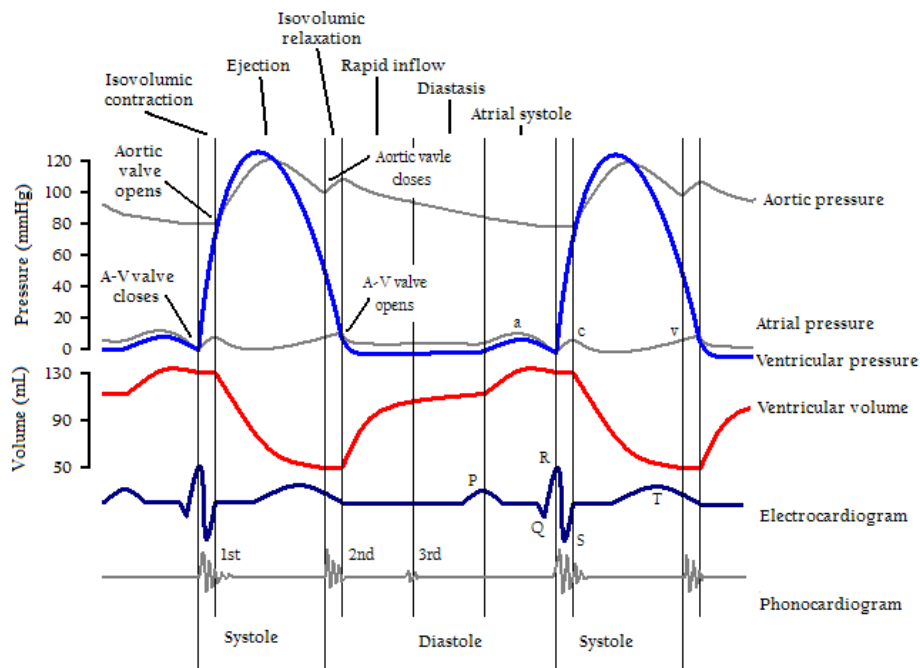


Figure 2 Heart Anatomy. Taken from [1]

## Cardiac Cycle

The cardiac cycle is a synchronized sequence of contractions and relaxations of the atria and ventricles during which major events occur, such as valves opening and closing and changes in blood flow and pressure. Each contraction and relaxation is referred to as systole and diastole, respectively. The next figure shows the events related to the cardiac cycle.



**Figure 3** Events and changes occurring during a cardiac cycle. Two complete cardiac cycles are shown. Taken from [2]

The figure's diagram starts at late ventricular diastole. At this stage, the AV valves are open and the ventricles near their maximum blood volume capacity. Atrial systole will then occur, pushing the blood through the AV valves, filling the ventricles even more, increasing their pressure and volume.

Next, as the ventricles begin to contract (ventricular systole), ventricular pressure (VP) rises above atrial pressure (AP), forcing the AV valves to shut. Since the semilunar valves are also closed, ventricular volume remains constant during this small period, known as Isovolumetric Contraction, causing a rapid increase in VP. When VP exceeds the pressure of the exit vessel (pulmonary artery and aorta for the

right and left heart, respectively) the semilunar valves will open, leading to the ejection of blood.

As the systole ends, the ventricular walls begin to relax (ventricular diastole) causing VP to drop drastically, falling below the exit vessel pressure, which causes the closure of semilunar valves. This period is referred to as Isovolumetric Relaxation because both semilunar and AV valves are closed, resulting in a constant ventricular volume and a further drop in VP. When VP falls below AP, AV valves open and blood flows into the ventricles, finally completing the cycle.

## 3.2. Heart Sounds

Heart sounds can be heard throughout the heart cycle and are caused by several cardiac events such as ventricular filling, blood flow and, most of all, valve movements. There are four main heart sounds, called S1, S2, S3 and S4. Normally, only two sounds are audible, S1 and S2.

- S1 – The first heart sound occurs right after ventricular systole, when both AV valves shut. Although only one sound is perceptible, two components are present, relative to the closure of each AV valve, and are known as T1 for the tricuspid valve and M1 for the mitral valve. As the left ventricle contracts first, the M1 component will occur earlier. This delay between M1-T1 has a major clinical importance and is considered physiological if under 30 ms. Above this value it may indicate a pathological split of S1 [3,4]. S1 has a typical duration of 100-200 ms [4,5] and its frequency spectrum is located mainly between 10-200 Hz [4,6,7].
- S2 – The second heart sound occurs at ventricular diastole and is caused by the closure of the semilunar valves. Just like S1, S2 has also two major components, relative to each valve: A2 for the aortic valve and P2 for the pulmonary valve. Since the aortic pressure is superior to the pulmonary pressure the aortic valve will close first, causing the A2 component to appear before P2. The delay can be affected by respiratory movements: during inhalation, as the chest expands, lung pressure drops and air flows

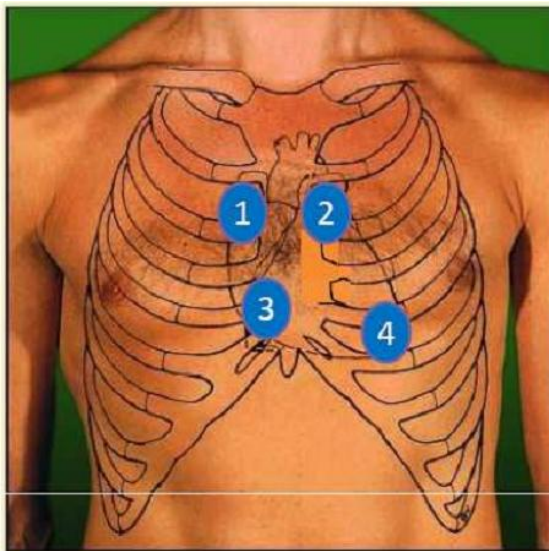


in. This process lowers the pulmonary pressure and the closure of the pulmonary valve will be delayed even further, increasing the split between A2 and P2. On the other hand, during exhalation, as lung pressure rises and forces the air out, the pulmonary pressure will increase, leading to an earlier closure of the pulmonary valve and lessening the split between A2 and P2. [3,4] Due to this fact, S2 maximum duration may vary considerably, while its minimum is approximately 80 ms. [8] The frequency components of S2 are typically higher than those of S1 and its spectrum is located mainly between 50-250 Hz [6,7].

- S3 – The third heart sound can be heard at diastole, moments after S2. It is believed it is caused by vibrations of the ventricular wall at the rapid filling period. The presence of S3 may be physiological in youths, young adults or athletes, while in elders it is usually related with heart failure and other pathologies. S3 is a low amplitude and low frequency sound whose spectrum is between 25-70 Hz. [3,4,9,10]
- S4 – The fourth heart sound occurs just before S1, at late ventricular diastole, and is caused by ventricular vibrations after atria contraction. It is also a low amplitude and low frequency sound and its presence is almost always pathologic. [3,4]
- Murmurs – Heart murmurs are high frequency sounds resulting from turbulent blood flow across narrow heart valves. The turbulent flow is caused by abnormalities such as constricted or leaking heart valves. These sounds are classified in 6 grades according to intensity, from 1 – faintest, to 6 – loudest murmur, and are usually associated with pathologic conditions like heart valve stenosis or insufficiency, for example. An “innocent murmur” is a murmur caused by physiological conditions and can be heard in children and athletes. [3,4]

### 3.3. Cardiac Auscultation

Cardiac auscultation is one of the oldest methods for heart function assessment, becoming a reliable technique after the invention of the stethoscope, by René Laennec in 1816 [4]. Auscultation is a non invasive, low cost method, which provides physicians rapid and fairly accurate information for heart diagnosis. On the other hand, interpretation of heart sounds is dependent on experience and skill, requiring years of practice.



**Figure 4** The four main auscultation sites: 1 – Aortic; 2 – Pulmonary; 3 – Tricuspid; 4 – Mitral. Taken from [5 p. 63].

As shown in Figure 4, there are four major auscultation sites, one for each heart valve, located where the respective heart sounds are heard with the highest amplitude:

- Aortic site (1) – between the second and third right intercostals spaces. Sounds originated from the aortic valve, such as S2 and murmurs, are best heard at this spot.
- Pulmonary site (2) – symmetric to the aortic site, between the second and third left intercostal spaces. Similarly to the aortic site, since S2 is

caused by the closure of both semilunar valves, this sound is also heard best at this location, as well as murmurs originating from the pulmonary valve.

- Tricuspid site (3) – located between the third to sixth intercostal spaces, it provides good audibility for S1, S2 and S3, when present, as well as other sounds originating from the tricuspid valve.
- Mitral site (4) – located between the fifth and sixth left intercostals spaces, at the apex of the heart. Just like the tricuspid site, both S1 and S2 can be heard, however, S1 is usually louder than S2. When present, S3 and S4 are also audible at this area.

The recording of heart sounds is known as Phonocardiogram (PCG) (Figure 5) and permits an objective and quantitative analysis to heart sounds, without depending on clinician's hearing capabilities and experience. However, the fact that different heart sounds have maximum amplitudes for different areas may pose problems when analyzing PCG signals as they are recorded from just one spot, leading to low intensity sounds.

In addition, several factors can contribute to low quality sounds: respiratory sounds, skin friction and ambient noise, muscular and bowel movements, fetal sounds in pregnant women and weak sounds for obese patients. [3,4,5]

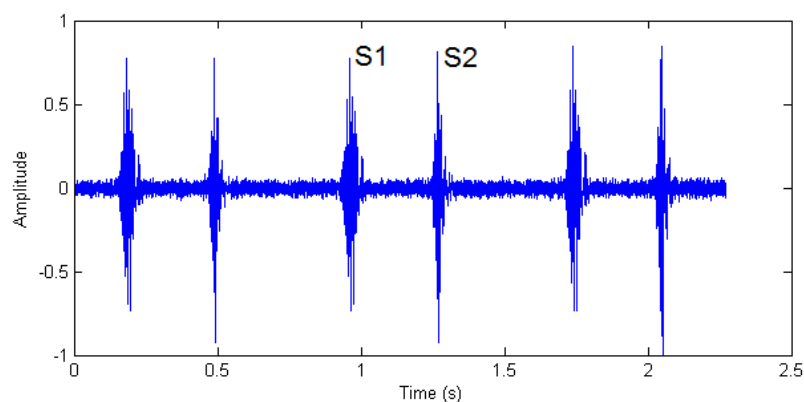


Figure 5 Phonocardiogram showing three normal cardiac cycles and the S1 and S2 heart sounds

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## 4. Related Works

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### 4.1. Background Problem

According to World Health Organization, CVDs are the leading cause of death Worldwide. In 2004, the estimated toll from CVDs was over 17 million, rising almost 2.4% from the 16.7 million deaths reported in 2002. This number is expected to hit 23.6 million by 2030. [1]

The economical burden of CVDs is tremendous: in the United States alone, the cost of CVDs was an estimated \$475.3 billion in 2009. [2] In Europe the situation is alarming as well, costing €192 billion in 2006. Direct healthcare costs account for 57% of that value and represent the cost of healthcare professionals, hospital fees and home nursing services. [3]

Several studies have demonstrated the efficiency of remote monitoring cardiac patients through auscultation [4,5] and the potential reduction of mortality rates, number of hospitalizations and the respective duration. Additionally, the use of automatic devices for diagnosis would improve results accuracy by providing an objective analysis independent from clinician's experience.

Remote monitoring not only improves patient's quality of life by avoiding hospitalizations, improving patient's recovery and allowing access to specialized medicine, but also by significantly reducing costs: according to [6], the adoption of remote monitoring technologies for congestive heart failure could save \$102.5 billion in the next 25 years, just in the United States alone.

Furthermore, the use of electronic stethoscopes would improve diagnosis accuracy by providing top quality acoustics and reducing the dependence on the physician's hearing skills.

## 4.2. State of the Art

### 4.2.1. ds32a Digital Stethoscope/Thinklabs® Phonocardiography Software

The ds32a Stethoscope is an electronic stethoscope by Thinklabs Medical. It features an amplify mode for auscultation sounds, an acoustic mode which simulates conventional stethoscopes, bell and diaphragm modes to listen to low and high frequency sounds and ambient noise filtering. It can also be connected to the iPhone, iPod and computers to record, listen or send the sounds to the clinician for an expert diagnosis. [7]



Figure 6 Thinklabs® ds32a Digital Electronic Stethoscope. Taken from [7]

Thinklabs Phonocardiography Software is an application for viewing and analyzing auscultation sounds. It was built on Audacity, an open source software program for digital audio editing and recording, and presents several features and sound processing capabilities:

- Import recordings from iPod and mp3 players;
- Visualize auscultation sounds waveforms
- Display spectrograms, Fourier Transform of sounds
- Sound filtering, amplification
- Free source code enables users to develop new tools for sound analysis

Despite the sound processing capabilities, the software provides only basic tools for sound analysis, lacking diagnosis features such as murmurs and heart sounds identification.

The digital stethoscope is available for \$249 while the software can be downloaded for free from the official site [7].

#### 4.2.2. 3M Littmann® Electronic Stethoscope Model 3200/Zargis® Cardioscan

The Littmann® Electronic Stethoscope Model 3200 is a high performance stethoscope designed especially for cardiology. It features proprietary ambient noise reduction, reportedly cancelling out 85% of background noise without affecting important body sounds; frictional noise reduction; sound recording and reproduction; bell and diaphragm modes for different frequency response to low and high frequencies and sound amplification.

With the purchase of the stethoscope, a software program for visualization of auscultation sounds is included – Zargis® StethAssist – to which the stethoscope can be connected via Bluetooth. The software can display sound waveforms in real-time, play recordings, saving patient information and send e-mail attachments. [8]



Figure 7 Zargis® Cardioscan interface and Littmann® Stethoscope Model 3200. Taken from [8]

Additionally, Zargis® Cardioscan can be acquired separately. It has the same capabilities of StethAssist plus it detects S1 and S2 heart sounds and identifies suspected systolic and diastolic murmurs. [9] The stethoscope is available for \$360 and Zargis® Cardioscan software for \$440.

### 4.2.3. Medtronic® CareLink Network

Medtronic® offers a varied range of devices for heart failure patients. The Medtronic® CareLink Network is an internet-based service which enables the remote monitoring of patients at home, by allowing the transmission of the information generated by the patient device to a healthcare provider. [10]



**Figure 8 CareLink Home Monitor. Taken from [10]**

Using CareLink Home Monitor, patients can send the information collected by the device over standard phone lines. The information is then stored in a secure database and the clinician can access it on a protected website for further analysis. Additionally, CareLink Network has an alert system via website, accessible by the clinician, providing alert notifications when a potential issue with a patient is detected.

This system improves efficiency on the daily care of risk patients, allowing them to be regularly monitored at home avoiding travels, reducing costs and improving patient's comfort and quality of life.



#### 4.2.4. CareTone® Telephonic Stethoscope

The CareTone® Stethoscope allows the transmission of high quality auscultation sounds over standard phone lines and digital lines. It is composed of two units, a sender and a receiver. The sending unit consists of a chest piece for the auscultation and headphones for examinations. The receiving unit contains headphones, volume control and a switch for bell or diaphragm modes for low and high frequencies, respectively.



Figure 9 CareTone® Stethoscope

The system was designed for heart, lung and bowel sounds and supports real-time transmission. One of the applications it is intended for is integration with videoconferencing and it is, reportedly, compatible with most telemedicine systems. [11]

Similar to the previous system, it allows a patient to be examined at home, with the same benefits as before. Furthermore, it allows different clinicians to actively participate in critical cases and have easy access to auscultation sounds when not at the same location as the patient, improving the precision of the diagnosis and the treatment.

### 4.2.5. Review

Although the first two systems are intended for healthcare professionals, they could potentially be used by patients since the recorded sounds can be stored and transferred to a computer, and later be sent to the clinician via internet. On the other hand, they could be quite expensive for the sole purpose of recording and storing auscultation sounds.

The third system presents interesting capabilities being quite simple to use and capable of generating alarms, although it has the inconvenience of always depending on a healthcare professional for the analysis of the clinical data collected.

Finally, the fourth system offers real time monitoring and interaction and can be integrated in video conferencing.

Of all the presented systems only Littmann® Stethoscope/Zargis® Cardioscan can provide a diagnosis, however, it depends on external hardware and is quite expensive. Moreover, the conducted research found no remote monitoring devices currently on the market with local processing capabilities, capable of providing a diagnosis.

The Look4MySounds platform covers almost all of the previous features, except the real-time monitoring and patient-clinician interaction, plus it offers diagnosis capabilities, while maintaining a low price.

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## 5. Look4MySounds

The objective of Look4MySounds project is to provide a quality healthcare service to patients who need constant medical observation, by allowing them to be monitored at home, significantly improving patient's quality of life while costs associated with travelling and hospitalizations are reduced.

### 5.1. Hardware specifications

The Look4MySounds system's prototype is composed of an audio board, an SD and MMC memory cards board, a microcontroller development board and a stethoscope:

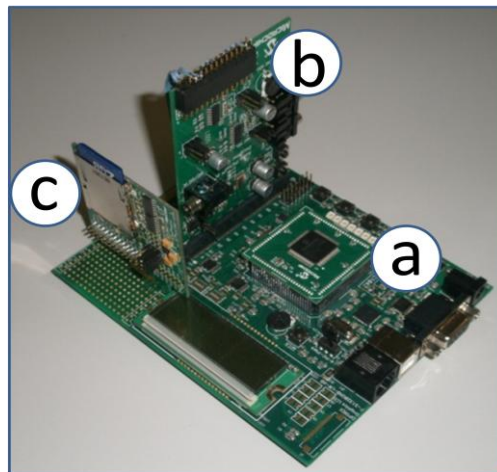


Figure 10 Look4MySounds prototype

- a) Explorer 16 Development Board – includes the microcontroller dsPIC33F and is responsible for the integration between all the boards. [1]
- b) Audio PICtail Plus Daughter Card – responsible for digitizing the sounds transmitted through the stethoscope. Permits various sampling rates of acquisition, ranging from 8 to 48 KHz. [2]

- c) PICtail Daughter Board for SD™ and MMC cards – contains the SD memory card reader. [3]

The stethoscope used was a Select model from Littmann® [4], which provides reliable and clear acoustics for auscultation. The microphone used to connect the stethoscope to the audio board was an electret microphone [Annex A] with frequency range 20 - 20000 Hz, ideal for heart sounds.



Figure 11 Stethoscope connected to the Look4MySounds prototype

## 5.2. System architecture

The system is intended to be easily operated by the patient or someone assisting and is powered by home power supply or battery. Auscultation sounds are locally processed and stored in a memory card. Next, the diagnosis and the sounds are transferred to a computer via RS-232 and sent to a healthcare professional by e-mail. The final objective is to implement GPRS to allow remote transferring of data.

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[Cited: August 26, 2010]

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# 6. Phonocardiography Signal Processing

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## 6.1. Wavelet Transform

The Wavelet Transform is a signal processing tool which provides time-frequency representation by decomposing a signal. The Wavelet Transform was developed to overcome one of the major limitations of Fourier Transform – the lack of time resolution.

### 6.1.1. Fourier Transform

The Fourier Analysis was introduced by Joseph Fourier in the early 19th century, suggesting functions could be decomposed into a sum of sines and cosines.

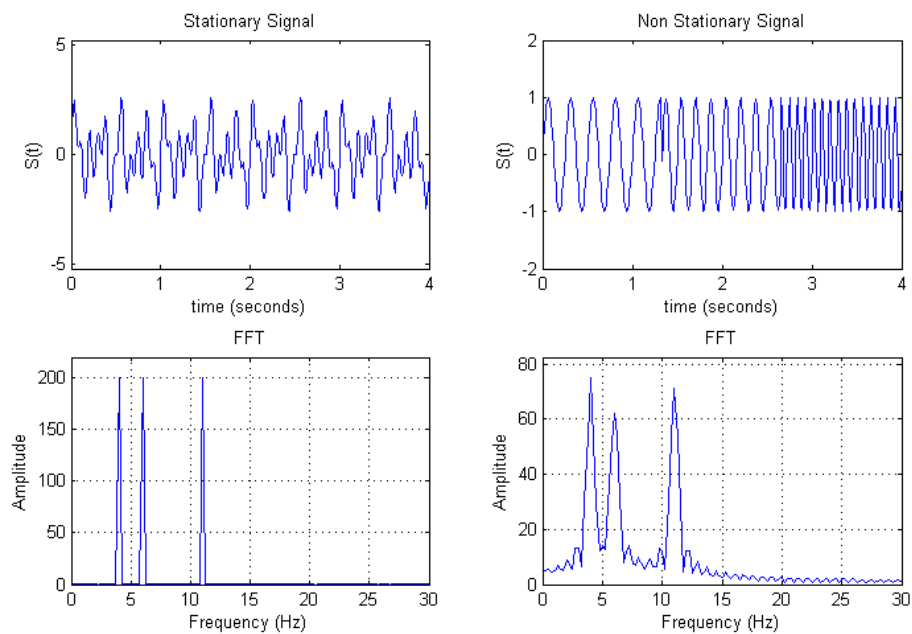
By decomposing a temporal signal in a sum of sinusoids we are representing it in the frequency domain which might reveal important details otherwise imperceptible in time domain. The tool used to obtain the frequency components of a signal is called Fourier Transform (FT) (Equation 6.1) and is one of the most used tools in signal processing.

$$X(f) = \int_{-\infty}^{+\infty} x(t)e^{-2i\pi ft} dt \quad (6.1)$$

Equation 6.1 defines the FT – it shows how the time-domain function  $x(t)$  can be transformed in the frequency domain function  $X(f)$ , where  $t$  and  $f$  stand for time and frequency, respectively. Conversely, it is possible to obtain  $X(f)$  through  $x(t)$  using the Inverse Fourier Transform (IFT) (Equation 6.2).

$$x(t) = \int_{-\infty}^{+\infty} X(f)e^{2i\pi ft} df \quad (6.2)$$

By applying the (FT) to a temporal signal to obtain its representation in frequency domain it is important to note that time information is lost. This means we know which frequency components exist in the signal but not which instants those frequencies are present at. This fact is very important when applying the FT to signals whose frequency components vary over time – called non stationary. On the other hand, signals whose frequency components are the same at all times are called stationary. The next figure illustrates this concept.



**Figure 12 Stationary and Non Stationary signals and the corresponding FT. Adapted from [1]**

Both signals from Figure 12 contain frequency components of 4, 6 and 11 Hz, however, they are completely different: the Stationary signal has those frequencies occurring at all times while in the Non Stationary signal they occur at different intervals. The corresponding FT for each of them, although not exactly the same, are very similar and show that both signals have three major spectral components of 4, 6 and 11 Hz.



This is the main limitation of FT: it has no time resolution, providing information about which frequencies are present in a signal but not when they occur. [2]

## 6.1.2. Continuous Wavelet Transform

The Continuous Wavelet Transform (CWT) is defined as:

$$\gamma(s, \tau) = \int x(t) \psi_{s,\tau}^*(t) dt \quad (6.3)$$

where \* is the complex conjugate and  $\psi_{s,\tau}$  is the wavelet function used to analyze the original signal. This function is derived from a general wavelet, called the mother wavelet (Equation 4), a predefined function which must satisfy several criteria.

$$\psi_{s,\tau}(t) = \frac{1}{\sqrt{s}} \psi\left(\frac{t-\tau}{s}\right) \quad (6.4)$$

The mother wavelet  $\psi$  is shifted in time using  $\tau$ , the translation parameter, and expanded or compressed using  $s$ , the scale or dilatation parameter, always greater than zero. This term is related to frequency: to analyze low frequencies of the signal, the mother wavelet is expanded ( $s>1$ ), and for high frequencies, the mother wavelet is compressed ( $s<1$ ). The term  $\gamma(s,\tau)$  is then calculated for all for values of  $s$  and  $\tau$ . This term can be seen as a measure of correlation between the signal and the wavelets obtained from the wavelet mother for all values of  $s$  and  $\tau$ , thus providing a time-frequency representation of the signal.

## 6.1.3. Discrete Wavelet Transform

The computation of CWT is done using discrete values of  $s$  and  $\tau$  with small increments, which takes a lot of computation power and provides redundant information. The Discrete Wavelet Transform (DWT) uses discrete

values for scale and translation, increased by powers of 2. This process ensures similar accuracy while significantly improving efficiency over CWT. [3,4]

### 6.1.3.1. Discrete Wavelet Transform Analysis

In 1988, Mallat developed an efficient algorithm implement the DWT, known as Subband Coding Algorithm, in which the signal in study is processed by a sequence of filters, so it can be decomposed in several frequency bands. The process starts by simultaneously filtering the signal with two half band low-pass and high-pass filters. The two signals obtained are then downsampled by a factor of 2 so each of them is half the size the original signal, as represented in the next figure. [2,3]

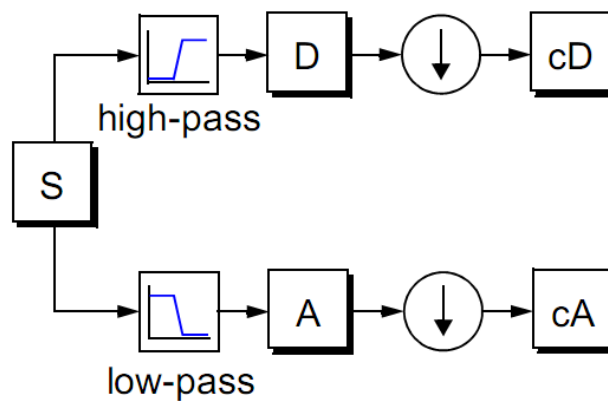


Figure 13 First step of DWT. Adapted from [3 p.1-25]

Figure 13 illustrates the first level of the decomposition, where two sets of coefficients are obtained: the output of the high-pass filter, followed by downsampling, is known as level 1 detail coefficients, abbreviated cD1. Similarly, for the low-pass filter, the output is called level 1 approximation coefficients, abbreviated as cA1.

Assuming the spectral components of the original signal range from 0-2000 Hz, the detail coefficients will correspond to the high frequency band of 1000 to 2000 Hz while the approximation coefficients correspond to the low frequency band of 0 to 1000 Hz.

The DWT is an iterative process and subsequent levels of the decomposition are achieved by applying the same step as in Figure 13 to the approximation coefficients from the previous, upper level (Figure 14).

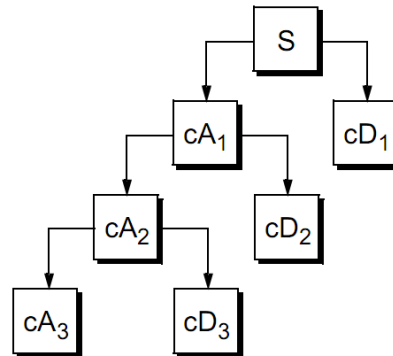


Figure 14 Wavelet decomposition tree. Taken from [3 p. 1-27]

On each level the same process is applied, so, considering the previous assumption, the second level approximation coefficients,  $cA_2$ , will cover the frequency range of 0-500 Hz, the detail coefficients for the same level,  $cD_2$ , will cover 500-1000 Hz and so on.

### Time-Frequency resolution

The prevalent spectral components in the original signal will appear with high amplitude at the respective coefficients which cover those same frequencies, conserving its temporal location. Despite this fact, time-resolution decreases as the level in which those frequencies appear is lower. This happens due to the fact that lower levels have fewer samples than higher levels for the same period, which translates in less time-resolution. That means the localization of frequencies present in lower levels will be less precise than those in higher levels. However, as the levels go deeper and time-resolution is lost, frequency resolution increases as the frequency band corresponding to the approximation and detail coefficients is smaller. [1]

### 6.1.3.2. Inverse Discrete Wavelet Transform

After a signal is decomposed using the DWT, it is possible to reconstruct it with the Inverse Discrete Wavelet Transform (IDWT). The reconstruction is done using the DWT coefficients and carries no loss of information.

The process, similar to the DWT, consists on upsampling the DWT coefficients followed by filtering. To upsample the coefficients, zeros are inserted between each sample. The obtained signal, twice the size of the original coefficients, is then filtered using a similar filter to the one used in DWT analysis for the respective coefficients. [3]

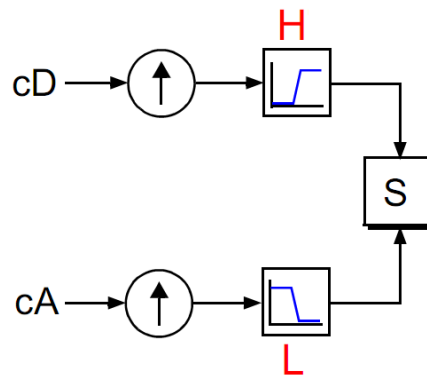


Figure 15 IDWT reconstruction. Adapted from [3 p. 1-30]

The reconstruction of the detail and approximation coefficients from certain level  $n$  originates the approximation coefficients from the level  $n-1$ . By reconstructing each level consecutively, the outputs from level 1 are finally combined to generate the original signal.

The most useful feature of reconstruction is that it is possible to reconstruct the approximations and details from the respective coefficients. Reconstructed approximations and details have the same size as the original size. Consequently, by adding the desired approximations and/or details from certain levels, it is possible to obtain a signal which is a representation of the original one for the frequency bands corresponding to the coefficients used for reconstruction.

## 6.2. Envelopes

Many segmentation algorithms for heart sounds rely on envelopes to detect sound lobes. An envelope is, as the name suggests, a smooth curve covering the whole signal. Although envelopes have less information than the signal itself, they facilitate the detection of sound lobes and the respective boundaries.

There are several methods to extract the signal envelope, being Homomorphic Filtering and the Average Shannon Energy two of the most used. [5,6,7]

### 6.2.1. Homomorphic filtering

Homomorphic Filtering is a technique used mostly in image processing, usually for quality enhancement or image correction. It consists on transforming the original signal, non-linearly, to a different domain. The new signal is then processed using suitable linear filtering and transformed back to the original domain. This process permits the elimination of unwanted components of the signal, otherwise impossible to do in the original domain.

#### 6.2.1.1. Homomorphic Envelopogram

This section describes how to obtain the envelope of a signal using Homomorphic Filtering.

Theoretically, a signal can be seen as a product of its frequency (FM) and amplitude (AM) components:

$$s(t) = f(t) \times a(t)$$

As the first step, we consider the logarithm of the modulus of the signal,

$$s'(t) = \ln|f(t) \times a(t)|$$

Since the logarithm is being applied, zeros in the signal must be first eliminated, so, a small, arbitrary value is added to the modulus, before applying the logarithm.

Considering the properties of logarithm and knowing the AM component is always positive we have:

$$\begin{aligned} s'(t) &= \ln|f(t) \times a(t)| \\ &= \ln|f(t)| + \ln a(t) \end{aligned}$$

The FM component has higher frequencies, thus, by applying an appropriate low-pass, linear filter  $L$  that component can be eliminated while the AM one is preserved:

$$\begin{aligned} s_f'(t) &= L[\ln|f(t)| + \ln a(t)] \\ &= L[\ln|f(t)|] + L[\ln a(t)] \\ &= L[\ln a(t)] \\ &\approx \ln a(t) \end{aligned}$$

Finally, the envelope, called Homomorphic Envelopogram, is calculated by exponentiation of the last term,

$$\textit{Envelope} = e^{\ln a(t)} = a(t)$$

The obtained envelope is shown in the next figure.

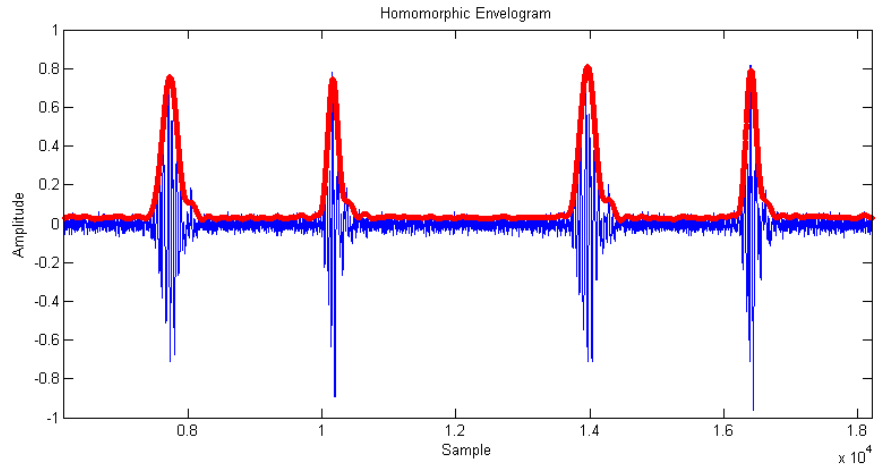


Figure 16 PCG signal (blue) and its Homomorphic Envelopogram (red)

### 6.2.2. Average Shannon Energy

The Average Shannon Energy is based on an operator, called Shannon Energy, defined as:

$$E_s = -s^2 \times \log (s^2)$$

where  $s$  is the original signal. By applying this operator to each sample of  $s$ , which must be first normalized to its absolute maximum, between -1 and 1, the Shannon Energy operator will affect the amplitude according to the following figure:

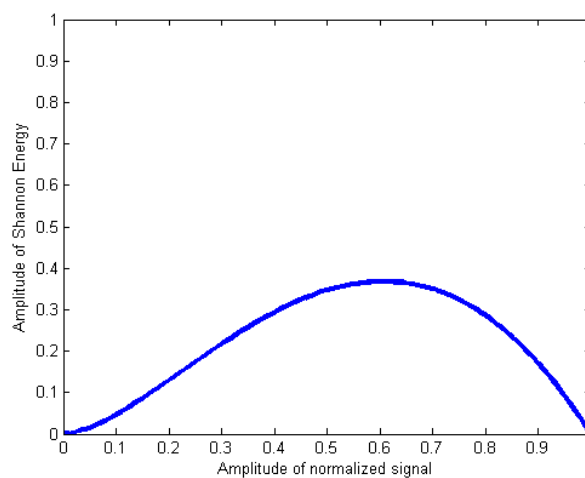


Figure 17 Shannon Energy amplitude versus Signal amplitude

As Figure 17 shows, the medium intensity samples of the original signal will be accentuated and the lower intensity samples will be attenuated more than the high intensity ones.

### 6.2.2.1. Shannon Envelope

The first step to calculate the Shannon Envelope consists on normalizing the signal between -1 and 1. Then, the Average Shannon Energy is calculated similarly to a moving average filter, using a 0.02 second window with 0.01 second overlapping according to:

$$E = -\frac{1}{N} \sum_{i=1}^N s_{norm}^2[n] \times \log(s_{norm}^2[n])$$

where  $N$  is the length of the corresponding window and  $s_{norm}$  is the normalized signal.

This operation will return an envelope of the signal, usually called Shannon Envelope, shown in the next figure.

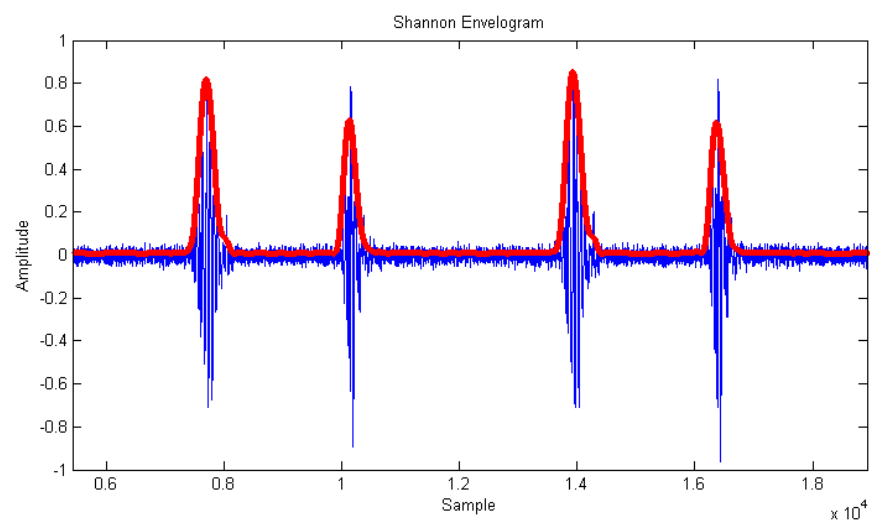


Figure 18 PCG signal (blue) and its Shannon Envelope (red)



### 6.2.3. Review

Both methods provide similar results, however, the Shannon Envelope is usually smoother than the Homomorphic Envelopogram, and might not detect low amplitude or short duration sounds. On the other hand, noise and artifacts with those characteristics might as well be eliminated. Consequently, the technique used must be chosen according to the signal in study.

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# 7. Algorithm Implementation

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This Chapter discusses the implementation of the algorithm for S1 and S2 heart sounds identification without any external reference. The development was done in Matlab 7.6.0 using an Intel Core 2 Duo (2.66 GHz).

The algorithm consists of three stages:

- **Pre-processing** – the purpose of this step is to eliminate noise and enhance heart sounds, making sound lobes easier to segment and identify.
- **Segmentation** – in this stage, the signal envelope is calculated, relevant sound lobes are segmented and artifacts and noise discarded according to duration and amplitude.
- **S1 and S2 identification** – previously segmented sound lobes are identified as S1 or S2. As stated in [1], this is one of the most difficult problems in heart sound analysis.

## 7.1. Pre-processing

High quality signals are essential for correct diagnosis. Unfortunately, the presence of noise in PCG signals is inevitable. Even when all background noise is minimized there are always intrinsic sounds impossible to avoid: respiratory sounds, muscular movements and so on. Therefore, the pre-processing stage is extremely important, ensuring elimination of noise and emphasizing relevant sounds.

### 7.1.1. Filtering

The first step in pre-processing is noise filtering. Several methods have been suggested, including DWT decomposition, where the signal is decomposed to a certain level, followed by reconstruction using the

coefficients corresponding to the frequency bands that contain the majority power of relevant heart sounds. [1,2]

The filtering process used consists on a band-pass filter applied to the raw PCG signal. Considering the frequency components of S1 and S2 heart sounds discussed in Chapter 3, the chosen frequency band for the filter was 20-250 Hz. This interval ensures the main components of those sounds are preserved, while high frequency noise is reduced as well as low frequency baseline fluctuations.

### 7.1.2. Shannon Energy

After the raw PCG signal is filtered, the obtained signal is normalized between -1 and 1, and the Shannon Energy operator is applied. As discussed in Chapter 6, the Shannon Energy emphasizes medium intensity samples while low intensity samples are attenuated. This process was found to improve segmentation results.

## 7.2. Segmentation

To segment relevant sounds, the signal envelope was extracted using Homomorphic Filtering with a cutoff frequency of 40 Hz for the low-pass filter. This value was obtained by experimental tuning. The calculated envelope is shown in the next figure.

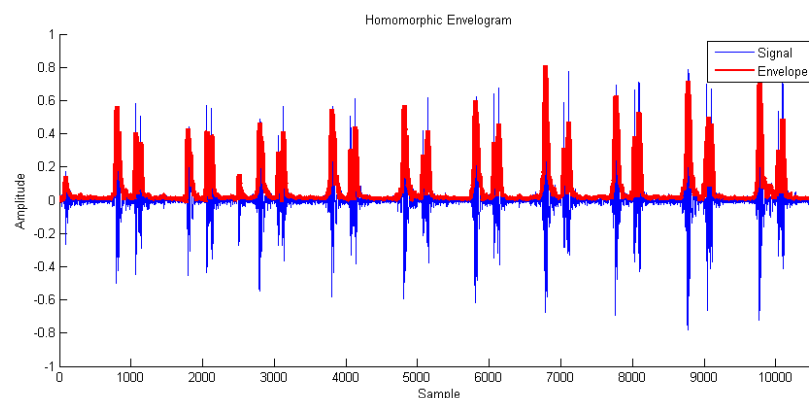


Figure 19 Homomorphic envelopogram of a PCG signal

After the envelope of the signal is obtained, a threshold value of 0.8 times the mean value of the envelope is set, and amplitudes above that level are considered relevant heart sounds. All sounds separated by less than 50 ms are considered the same sound, as suggested in [1,3], while those with duration inferior to this value are discarded. The 50 ms value was chosen according to maximum interval for splitted sounds [4] as well as minimum duration for heart sounds. The results of this stage are shown in the next figure.

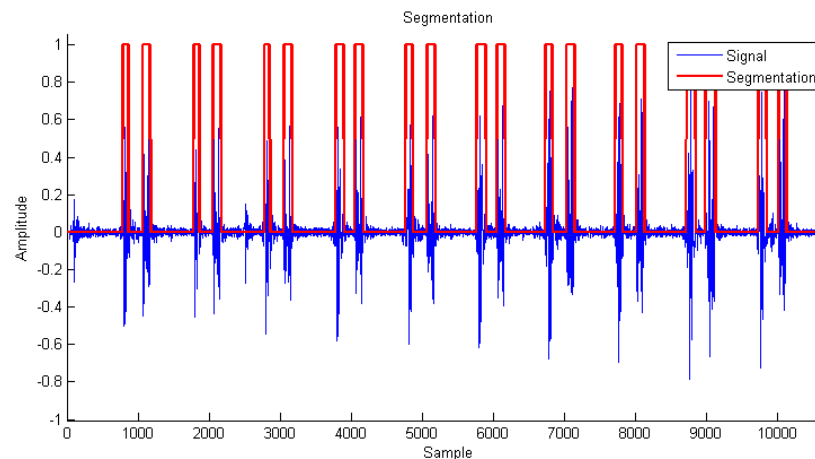


Figure 20 Final results of the Segmentation stage

### 7.3. S1 and S2 identification

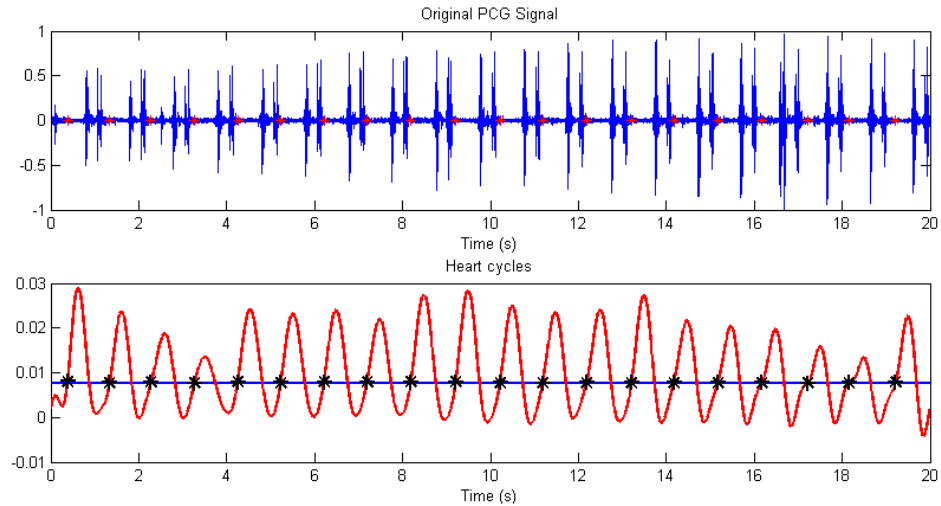
Once the sounds are segmented, they need to be identified as S1 or S2. Initially, the first approach towards this problem was using DWT decomposition. According to [1,5], the frequency of valve vibrations is proportional to the pressure difference across that valve, causing the S2 sound to have higher frequency components than S1, as the pressure difference across the aortic valve is greater. This property was used to identify S2 heart sounds using DWT coefficients for high frequencies, however, results were not the expected and the method was abandoned.

A new identification method was explored, based on the fact that, for normal heart rates, the diastolic period is longer than the systolic period. [3,6] Consequently, it is assumed the longest interval between two sounds corresponds to the diastolic

period and the immediate sound following that interval corresponds to an S1 while the immediate sound preceding it corresponds to an S2.

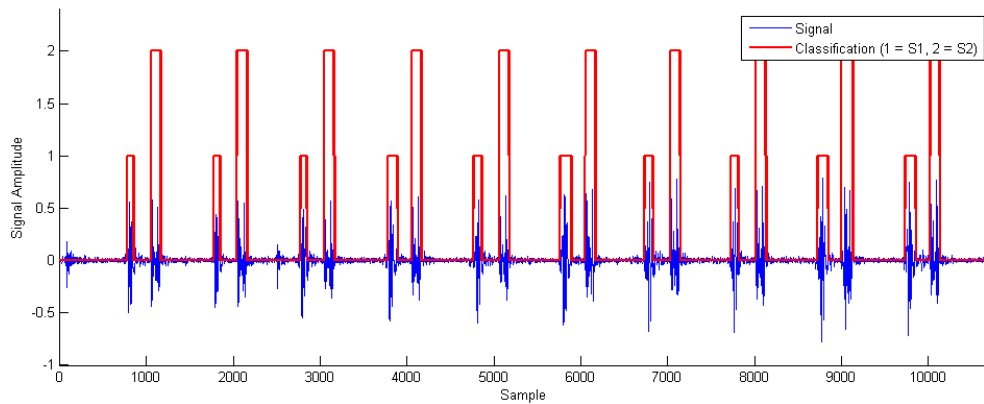
After the first S1 and S2 are found, adjacent sounds are classified using the cardiac cycle duration. To determine the cardiac cycle duration, heart rate is first calculated according to the following method:

- a) The absolute maximum of the signal is picked. It is assumed this point belongs to an S1 or S2 heart sound. By selecting all samples less than 60 ms from that point a 120 ms segment is obtained. The 120 ms value was chosen taking into account the mean duration of S1 and S2 heart sounds.
- b) Considering the difference between S1 and S2 waveforms, by cross-correlating the segment (which corresponds to an S1 or S2) with the whole signal, a new one is obtained, with peaks where the same type of sound of the segment is repeated.
- c) To the new signal, a procedure similar to the Pan-Tompkins algorithm for QRS detection in ECG signals [7] is applied: a band-pass filter for 5-20 Hz, differentiation, squaring and a moving average filter with a 0.3 sec window. Finally, homomorphic filtering is used, with a cutoff frequency of 2 Hz for the low-pass filter, resulting in a sinusoid-like waveform matching each cardiac cycle, as shown in Figure 21. Using a threshold value, the mean period is determined, which is considered to correspond to the cardiac cycle duration of the original PCG signal.



**Figure 21 Heart rate calculation**

After the heart rate is obtained, the mean period of the cardiac cycle is calculated and each S1 and S2 heart sound is classified relatively to the adjacent same type of sound, starting at the first S1 and S2 detected previously.



**Figure 22 Results for S1 and S2 identification**

Finally, all samples segmented as relevant sound in the Segmentation stage not classified as S1 or S2 are discarded.

## 7.4. Algorithm schematic

The next image (Figure 23) shows a summarized schematic of the developed algorithm.

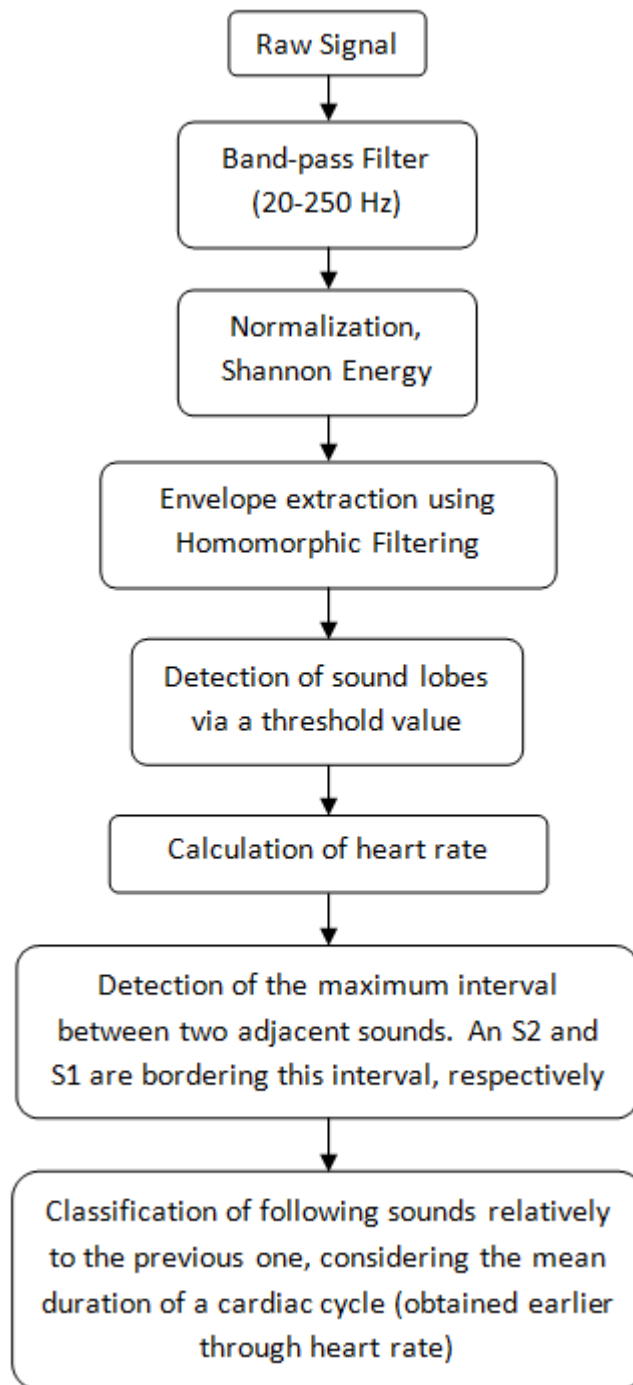


Figure 23 Algorithm Schematic

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## 8. Experimental Results

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The described algorithm was tested with two groups of data:

- A. Non pathologic sounds recorded by the student. 17 sounds were recorded using the stethoscope described in chapter 5 connected directly to a laptop computer using Matlab 7.6.0. All sounds were non pathologic and recorded between the aortic and mitral auscultation sites which offered the best sound quality for both S1 and S2 heart sounds. The recordings were done in a quiet room.
  
- B. Sounds collected online from two free databases, <http://www.medstudents.com.br/cardio/heartsounds/heartsou.htm> and <http://www.egeneralmedical.com/listohearmur.html>, consisting of 66 pathologic sounds and 4 non pathologic.

Additionally, pathologic sounds were recorded at the Cardiology Department of Hospital dos Covões, however, they were corrupted by noise and artifacts due to several reasons: background noise, weak heart sounds and respiratory sounds. Furthermore, the stethoscope and microphone used provided very low amplitude sounds. As a result, this data set was not considered for testing.

To assess the algorithm performance, sensitivity (SE) and specificity (SP) are considered. To calculate these parameters three variables are used: true positives (TP), false positives (FP), false negatives (FN). A TP is considered when a sound is correctly identified; a FP is considered when a sound is incorrectly detected; a FN is considered when a sound is not detected. The calculation of SP uses a slightly different expression from the regular one, which included the variable true negatives (TN) – correct classification of no event. Since there are no means to quantify TN in S1 and S2 detection, SE and SP, as in [1,2], are defined as:

$$\text{Sensitivity} = \frac{TP}{TP+FN} \quad (8.1)$$

$$\text{Specificity} = \frac{TP}{TP+FP} \quad (8.2)$$

A GUI was developed to record heart sounds and testing of the algorithm [Annex B]. The achieved results are presented in Table 1. The total number of cardiac cycles for groups A and B were 335 and 390, respectively. The complete classification of a 75 sec heart sound, with a sampling rate of 8012 Hz, took an average of 2.95 sec.

Group	Sound	Sensitivity	Specificity
A	S1	98.8%	97.9%
	S2	98.8%	98.2%
B	S1	86.9%	86.4%
	S2	94.8%	89.7%

**Table 2 Results for S1 and S2 identification**

## 8.1. Discussion

The algorithm performs very well for non pathologic sounds as shown by the results for group A, which were considerably better than those of group B. This is caused by the typical murmurs and extra heart sounds present in pathological sounds leading to incorrect classification. A few of the issues caused by these events have been identified:

- Loud diastolic murmurs were often segmented as relevant sounds, causing the systolic period to be the largest interval between two sounds present in the signal. Consequently, the first S1 and S2 were incorrectly identified and the classification process failed.

- Loud systolic murmurs would usually be segmented together with S1 and S2 into a single sound leading to incorrect classification.

## References

- [1] D. Kumar, P. Carvalho, M. Antunes, J. Henriques, L. Eugénio, R. Schmidt, J. Habetha, *Detection of S1 and S2 Heart Sounds by High Frequency Signatures*, 2006
- [2] D. Gill, N. Intrator, N. Gavriely, *A Probabilistic Model for Phonocardiograms Segmentation Based on Homomorphic Filtering*, 2006

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## 9. Conclusions and future work

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### 9.1. Conclusions

The initial planning of this work was to develop an algorithm for the automatic detection of pathologies in heart sounds signals. With the growing complexity of the problem it was decided that the best approach would be to first develop an algorithm for the identification of S1 and S2 heart sounds. In this perspective, the objective of this work was accomplished.

Regarding the stethoscope and microphone used to record the sounds, it provided very low amplitude signals. The non-pathologic sounds recorded by the student were collected from friends and family members, who were asked to be completely still and, sometimes, in apnea. During the recording of pathological sounds at Hospital dos Covões, the student had no control over the actions of the patients or background noise. This was the main reason between the sound quality of both non-pathologic and pathologic sounds.

One of the main weaknesses of the proposed algorithm is that it relies too much on diastolic period for the detection of the first S1 and S2. Although the diastolic period is almost always the longest interval between two sounds, in a few cases, as presented in Chapter 8, it doesn't apply. If the detection method for the first S1 and S2 doesn't succeed, the whole classification process will fail. Nevertheless, the obtained results were surprisingly better than expected, especially for pathological sounds.

The correct identification of S1 and S2 lays ground for the detection of pathologies, since by having S1 and S2 segmented, both the diastolic and systolic period can be analyzed for the presence of murmurs and extra heart sounds, indicating possible pathologies.

## 9.2. Future Work

One of the issues to overcome is sudden changes on heart rate. Although the analyzed signals are only a few seconds long, heart rate can rapidly change. The first improvement to the algorithm could be an adapting value for the heart cycle duration, which is calculated in the S1 and S2 identification stage and used to detect all S1 and S2 after the first ones. This means the value for the heart rate duration used to detect successive sounds would adjust according to the heart rate of the last few sounds detected.

The next step is to implement a method for the detection of pathologies. As said before, by having all S1 and S2 detected, most elements which characterize pathologies can be easily studied, including the diastolic and systolic periods for murmurs and the duration of S1 and S2 for pathological splits.

Concerning the high background noise found in data set B (pathologic sounds) the main cause was the assembly of the stethoscope and microphone. Although the stethoscope provides good quality sounds and the frequency response of the microphone is ideal for heart sounds, the microphone was simply inserted into the stethoscope tubing, which possibly caused air leaks, leading to the low amplitude sounds and background noise recorded. The use of an electronic stethoscope would potentially solve this problem.

The final step is to convert the developed Matlab code to C and implement the algorithm in the microcontroller. Clinical trials must follow to study the market viability.

# Annexes

## Annex A - Microphone

### SPECIFICATION FOR APPROVAL

### 承認書

Description : **Electret Condenser Microphone**

Kingstate Part No. : **KECG2742PBL-A**

Customer's Model No. :

Specification No. : **MKD-7161**

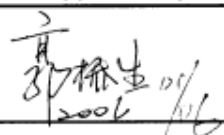
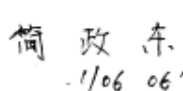
Number Of The Edition : **1.2**

CUSTOMER'S APPROVED SIGNATURE		

志豐電子股份有限公司 KINGSTATE ELECTRONICS CORP.



Address : 10F, No. 69-11, Sec. 2, Chung Cheng E. Rd., Tamshui County, Taipei Hsien, Taiwan, R.O.C.  
 International sales dept.: TEL:886-2-2809-5651 FAX:886-2-2809-7151  
 Domestic sales dept.: TEL:886-2-2809-0668 FAX:886-2-28096748  
<http://www.kingstate.com.tw>

Approved by	Checked by	Issued by
 郭標生 2006/06	 簡政东 1/06 06'	<b>Qin</b> <b>01/05/06'</b>

**A. SCOPE 範疇**

This specification applies elect ret condenser microphone, **KECG2742PBL-A**  
 此規格書適用於電容式麥克風 **KECG2742PBL-A**

**B. SPECIFICATION 規格**

No.	Item	Symbol	Unit	Specification	Condition
1	Directivity 方向性			Omnidirectional 全指向	
2	Sensitivity 靈敏度	S	dB	-42 ± 3	f=1KHz, 1Pa 0dB=1V/Pa
3	Standard operating voltage 標準操作電壓	Vs	V D.C.	2	
4	Output impedance 輸出阻抗	Zout	KΩ	2.2	f=1KHZ, 1Pa
5	Max operating voltage 最大操作電壓		V D.C.	10	
6	Sensitivity reduction 減電壓靈敏度	ΔS-Vs	dB	-3	f=1KHz, 1Pa Vs=2.0V D.C. to 1.5V D.C.
7	Frequency 頻率	f	Hz	20 ~ 20,000	
8	Max. current consumption 最大耗電流	Ibss	mA	0.5	Vs=2.0V D.C. RL=2.2KΩ
9	Signal to noise ration 訊號對雜音比	S/N	dBA	58	f=1KHz, 1Pa Aweighted
10	Operation temp. 操作溫度		°C	-20 ~ + 70	
11	Storage temp. 儲存溫度		°C	-20 ~ + 70	
12	Dimension 尺寸		mm	φ 6.0 x 2.7	See appearance drawing 請參照外觀尺寸圖
13	Weight (MAX) 重量		gram	0.17	
14	Material 材質			AL 鋁	
15	Terminal 端子			Pin type (鍍化金/Plating Au)	See appearance drawing 請參照外觀尺寸圖
16	Environmental Protection Regulation 環保法規			RoHS	

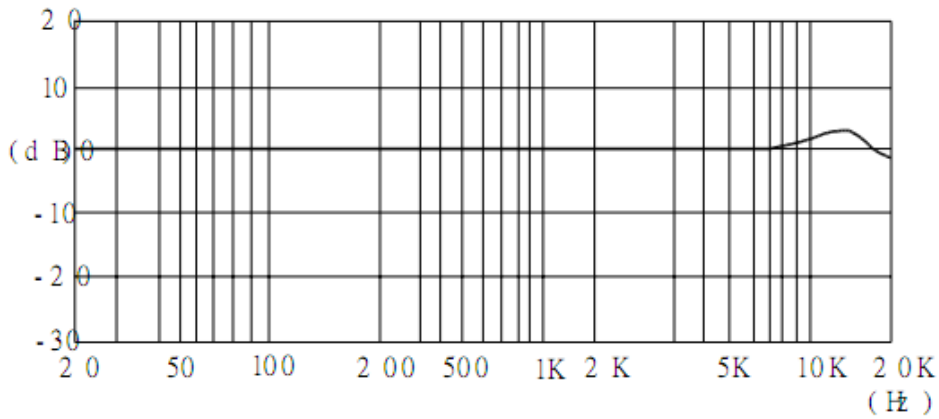
We use "Pascal (Pa)" indication of sensitivity as per the recommendation of I.E.C. (International Electro technical Commission). The Sensitivity of "Pa" will increase 20dB comparing with "ubar" indication.

Example: -60dB (0dB=1V/ubar) = -40dB (1V/Pa)

依I.E.C. (國際電子協會)建議,以"Pa"為靈敏度標示單位比"ubar" 靈敏度標示單位增加20dB

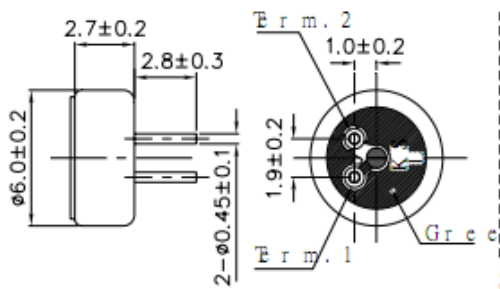
例如: -60dB (0dB=1V/ubar) = -40dB (1V/Pa)

C. TYPICAL FREQUENCY RESPONSE CURVE 頻率響應曲線



D. APPEARANCE DRAWING

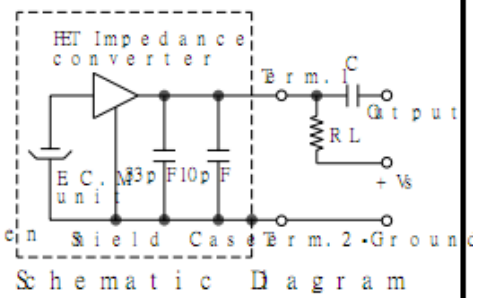
外觀尺寸圖



Unit: mm

E. MEASUREMENT CIRCUIT

測量線路



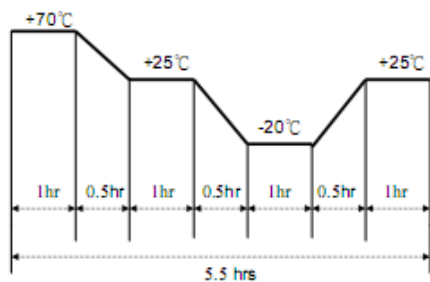
RL=2.2KΩ



**F. MECHANICAL CHARACTERISTICS 機械特性**

No.	Item	Test condition	Evaluation standard
1	Soldering Heat Resistance 焊錫耐熱性	Lead terminal are immersed in soldering bath of +270±5°C for 2±0.5 second. 端子部份浸入以+ 270±5°C 焊錫槽中 2±0.5 秒	No interference in operation 操作無任何不良
2	Terminal Mechanical Strength 端子強度	Apply the terminal with 4.9 N (0.5 kg) strength for 30 second 對端子施以 4.9 N (0.5 kg) 力量 30 秒	No damage and cutting off 端子不鬆動, 不脫落
3	Vibration Test 振動測試	The part shall be measured after being applied vibration of amplitude of 1.5mm with 10 to 55hz band of vibration frequency to each of 3 per-pendicular directions for 2 hours. 振動週波數 10~55HZ、全振幅 1.5mm 於 X、Y、Z 3 個方向, 各 2 小時	After any tests, the sensitivity to be within ±3dB
4	Drop Test 落下測試	The part unit without packaged must be subjected to each 3 drops at three axes from the height of 1 meter to 20 mm thick wooden board. 未包裝麥克風單體從 1 公尺高處, X、Y、Z 3 個方向, 各 3 回, 落於 20mm 厚木板上。	測試其靈敏度須與原測試值的差異不超過±3dB.

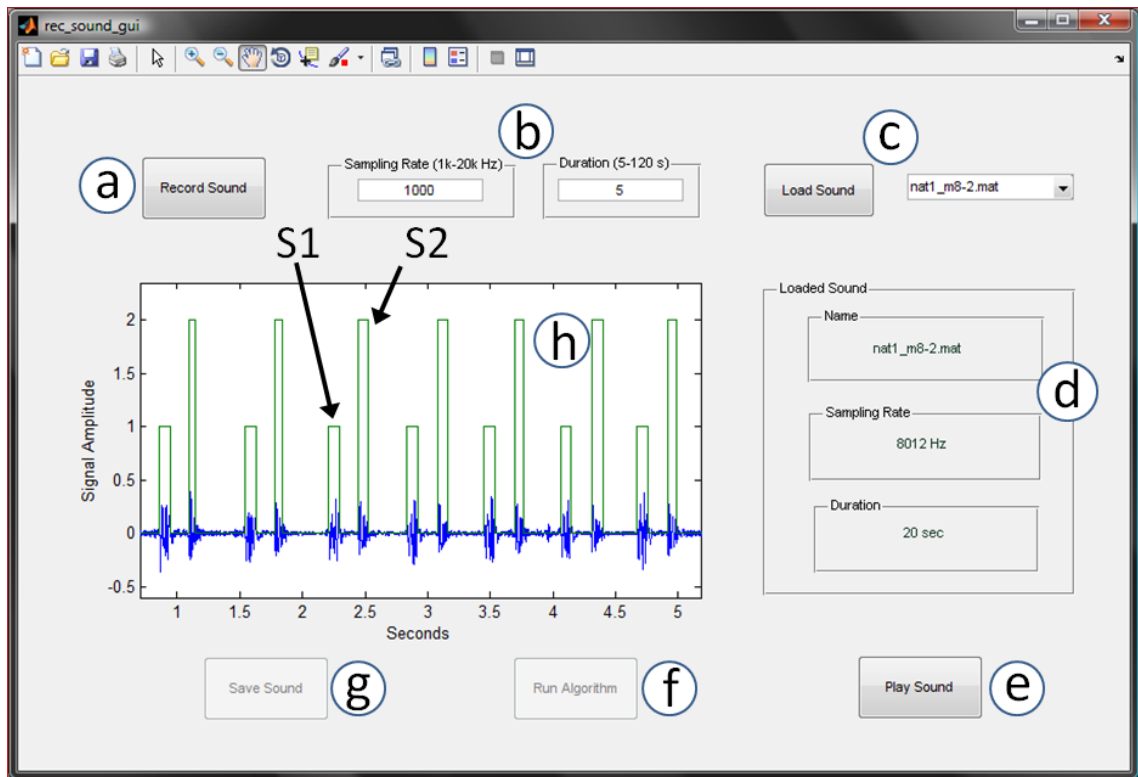
**G. ENVIRONMENTAL TEST 環境試驗**

No.	Item	Test conditions	Evaluation standard
1	High temp. test 高溫測試	After being placed in a chamber at +70°C for 72 hours. 置於+ 70°C 環境中 72 小時	After any tests, the sensitivity to be within ±3dB of initial sensitivity after 6 hours of conditioning at +25°C 經測試後麥克風, 須靜置於 +25°C (室溫) 環境中 6 小時後, 測試其靈敏度須與原測試值的差異不超過±3dB.
2	Low temp. test 低溫測試	After being placed in a chamber at -20°C for 72 hours. 置於-20°C 環境中 72 小時	
3	Humidity test 相對濕度測試	After being placed in a chamber at +40°C and 90±5% relative humidity for 240 hours. 置於+ 40°C, 相對濕度環境中 240 小時	
4	Temp. cycle test 溫度循環測試	The part shall be subjected to 10 cycles. One cycle shall be consist of: 單體承受溫度循環測試 10 次, 其循環內容如圖示: 	

**TEST CONDITION.**

Standard Test Condition : a) Temperature : +5 ~ +35°C b) Humidity : 45-85% c) Pressure : 860-1060mbar  
 一般測試條件 : a) 溫度 : +5 ~ +35°C b) 濕度 : 45-85% c) 氣壓 : 860-1060mbar  
 Judgment Test Condition : a) Temperature : +25 ± 2°C b) Humidity : 60-70% c) Pressure : 860-1060mbar  
 爭議時測試條件 : a) 溫度 : +25 ± 2°C b) 濕度 : 60-70% c) 氣壓 : 860-1060mbar

## Annex B – Graphical User Interface



Legend:

- a) Start recording
- b) Choose recording's sampling rate and duration
- c) Load existing sound
- d) Display of the parameters of the most recent recorded or loaded sound
- e) Play sound
- f) Run the classification algorithm
- g) Save sound
- h) Sound waveform (blue) and its classification (green; value 1 = S1, value 2 = S2)