TAMPEREEN TEKNILLINEN YLIOPISTO
TAMPERE UNIVERSITY OF TECHNOLOGY

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# AN ECG BRACELET WITH COMMUNICATION SYSTEM

Master of Science Thesis

Examiners:
Professor Jari Viik
Associate Professor Juha Nousiainen

Examiners and topic approved by the Council of the Faculty of Natural Sciences on April 9th 2014

## ABSTRACT

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Quality of life of patients needing repeated ECG measurements is threatened by the need of several hospital visits, cost inefficiency of having ECG measurements at hospitals and discomfort caused by many ECG leads. For such patients an ECG system with continuous measurements gives profound insight into the heart's condition. Therefore a low cost, lead I ECG measurement device with a universal communication system would benefit the patient significantly. In this thesis, a low cost electrocardiogram bracelet with communication system using a microphone jack was designed and developed from textile electrodes and electronic components.

Different variations and types of textile electrodes were compared and inspected in order to find the most suitable one for this biomedical application. The criteria for selecting the final textile electrode were low power spectral density, root mean square noise and electrical bioimpedance. The bracelet itself is made out of textile electrodes, with connectors to the communication system which will enable any computer or device with a sound port to detect, show and save the ECG measurements. This type of system enables for the equipment to be used in any place at the patients' convenience as long as he has access to a sound recording device with a microphone port.

To confirm the functionality and the quality of the ECG waveform recorded by the manufactured device, it was necessary to have a testing session to observe the performance of the device and communication system. After the initial measurements with laboratory equipment, functionality of the constructed system was tested with a BIOPAC MP36, after which the signal was recorded with a microphone jack.

The ECG bracelet obtained consistently good quality ECG, which leads us to see a future in this line of medical instrumentation.

# PREFACE

The idea for a non-clinical and easy to use ECG electrode and system was concocted while conversing about continuing with my bachelors' thesis regarding the addition of biomeasurement capabilities to an electrical bed that would transform into a wheelchair. With time this idea evolved and matured into the thesis that resides here. ECG was first developed due its importance in the medical world and its simplicity to measure. The idea also had helped to evolve which came from Professor Jari Vikk and Associate Professor Juha Nousiainen, whose help was immeasurable.

Time alone is not enough for an idea to happen, nor do they happen by chance. I thank my parents whose love and encouragement for the past 26 years that has led to this achievement. At the end of the day this achievement along with the rest would not have existed if not for them, so I will dedicate this thesis to the memory of my father and to the endless support from my mother.

This thesis couldn't have come into fruition without my grandparents' unfaltering help towards my education throughout my life. Last but not least I also want to thank my friends that have helped and encouraged me during my masters until the very end. I would specifically like to thank Defne Us for all her feedback, support and help in writing this thesis.

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# LIST OF ABBREVIATIONS

| | |
|---|---|
| A | Exponent parameter. |
| $\omega$ | Angular frequency. |
| $\tau$ | Time constant. |
| ADC | Analogue to digital converter. |
| Ag/AgCl | Silver silver-chloride. |
| ASIC | Application specific integrated circuit. |
| A-V | Atrioventricular. |
| CATIA | Computer aided three-dimensional interactive application. |
| CMRR | Common mode rejection ratio |
| EBI | Electrical bioimpedance. |
| ECG | Electrocardiogram. |
| $f_c$ | Cut-off frequency. |
| FPGA | Field-programmable gate array. |
| IOS | IPhone operating system. |
| ISO | International organization of standardization. |
| LA | Left arm. |
| LL | Left leg. |
| mmHg | Millimetre of mercury. |
| op amp | Operational amplifiers. |
| PSD | Power spectral density. |
| PET | Polyethylene terephthalate. |
| $R_0$ | Static resistor. |
| $R_\infty$ | Ideal resistor. |
| RA | Right arm. |
| RC | Resistor-capacitor. |

| | |
|---|---|
| REACH | Registration, evaluations, authorization and restriction of chemicals. |
| RL | Right leg. |
| RMS | Root mean square. |
| RoHS | Restriction of hazardous substances directive. |
| S-A | Sinoatrial. |
| TiN | Titanium nitride. |
| WHO | World Health Organization. |
| $Z_{cole}$ | Cole parameter. |

# 1 INTRODUCTION

It is imperative to remark how important cardiovascular diseases are, for example in a recent study published on 2013 by WHO, it was stated that in 2011 3 out of every 10 deaths, nearly 17 million people, died due to cardiovascular disease. [1] In the same study they published a comparison showing the major causes of death compared to the income level of the countries in which these deaths happen (Figure 1). This study indicates not only the importance of detecting and preventing heart diseases in countries of all income levels, but also the vulnerability of any human being to a heart disease, which is usually considered as unforeseeable.

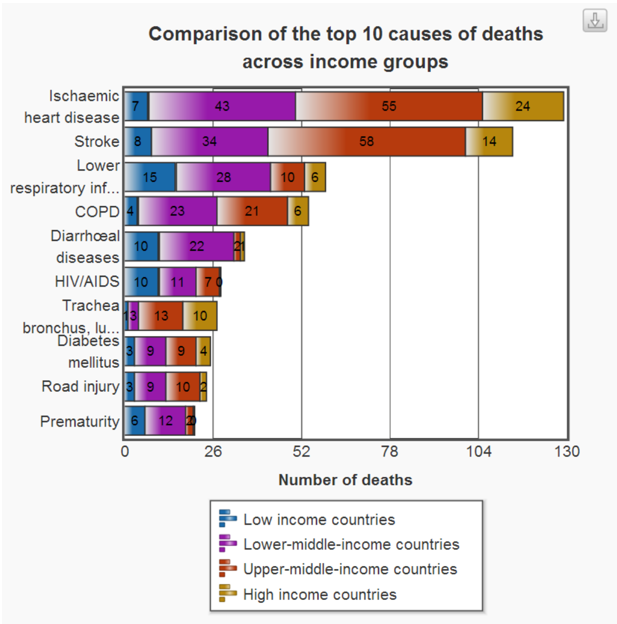



**Figure 1** Comparison of the top 10 causes of deaths across income groups [1]

However recent studies suggest that by means of using ECG, the risk of death in patients can be predicted. The study was performed in regards to patients with previous cardiac events and whose ECG was recorded shortly after, while they were in recovery. Collecting long term ECG data along with an estimate of activity levels, may give physicians a better estimate of the individual's cardiac health. Periodic ECG checks are quite helpful since there is a 77% chance of being identified as a high or low risk patient. [2]

During recent years, there has been an intensification of work towards maintaining quality of care while reducing overall costs. [3] One way in order to reduce these costs is by means of providing systems to monitor the individual outside of the hospital's facilities. Physicians could monitor individuals recovering from a cardiac attack, those at risk of one and those experiencing cardiac discomfort, while they perform their daily routine in their households.

Traditional electrodes present drawbacks when used in long term applications. Some of these drawbacks are allergic reaction to the electrolytic gel, not being able to be used for extended periods of time and the repeated need to be carefully placed. Textile electrodes, however, do not present allergies, they are reusable and the proper design of the garment ensures that it will be correctly positioned. They have given the possibility of making comfortable prolonged recordings of cardiovascular measurements. Another drawback introduced by traditional electrodes is various sources of interference during measuring; these are accentuated while using textile electrodes. Making it of grave importance to properly study and compare these types of electrodes.

The objective of this thesis is to design and manufacture a working prototype of an electrocardiogram (ECG) bracelet using textile electrodes with a special communication interface, for observing noticeable changes in the ECG waveform. In this chapter we will present the motivation for the purpose of the project. In the second chapter we will introduce some of the main concepts that will be dealt with during the thesis in order to give some background information. Continuously in the third chapter we will proceed to the work done and to explain the procedure in which the ECG bracelet was designed, manufactured and tested, along with the communication system. Then we will proceed in order to present and analyze the results, we will also discuss the findings in the fourth chapter, and finally we will present our conclusions in the fifth chapter, where we will also talk in regards to the future and probable implementations of this device.

# 2 THEORETICAL BACKGROUND

## 2.1 Electrocardiogram Measurement.

Since its discovery, over 100 years ago, electrocardiogram (ECG) has become a routine and an important part of any complete medical evaluation and has been used continuously over this time as a diagnostic test. [4] Most of the damages caused to the heart's tissue can be perceived by the ECG due to the fact that electricity is conducted through the heart muscle, also known as myocardium. The information obtained from the ECG is represented as the ECG waveform. This waveform provides us with information about the electrical activity associated to different aspects of the heartbeat; making it a very important tool to assess an individual's heart health and cardiac rhythm.

For a successful ECG measurement system design, it is important to understand the basic foundations of the electrocardiogram measurements. The often referred as a "sinus" rhythm is the normal cardiac rhythm; it is referred as sinus due to the fact that it originates in the sinoatrial node (S-A node). The heart beat is triggered when cells deemed as "pacemaker cells" found in the S-A node are stimulated. The disturbances of the normal sinus rhythm are commonly known as arrhythmias. The normal behaviour of the "pacemaker cells" is to generate action potentials at 60-100 beats per minute, when resting. The action potential later depolarizes the tissue before disseminating throughout the myocardium in a predefined way, triggering the tissue to contract. Then after leaving the S-A node, the action potential depolarizes both of the atrial chambers of the heart, from there it moves down to the atrioventricular node (A-V node), this being a node that is found between the atrial and ventricular chambers. The A-V node acts as a backup pacemaker, however, it acts at a lower rate of 40-60 beats per minute. This node acts as a delayer for the electrical impulses, slowing them down by 120 ms giving the atria enough time to pump their blood into the ventricular chambers, before they contract. Then the action potential leaves the A-V node and continues down to the area known as "HIS" bundle, from there it goes to the bundle branches to the left and to the right. Finally the ventricles are depolarized by the conductive fibres named Purkinje fibres. Finally after the repolarization and contraction of the ventricles, the cycle starts again. The ECG waveform is a recording of the superposition of these different action potentials, as can be seen in Figure 2. [5]

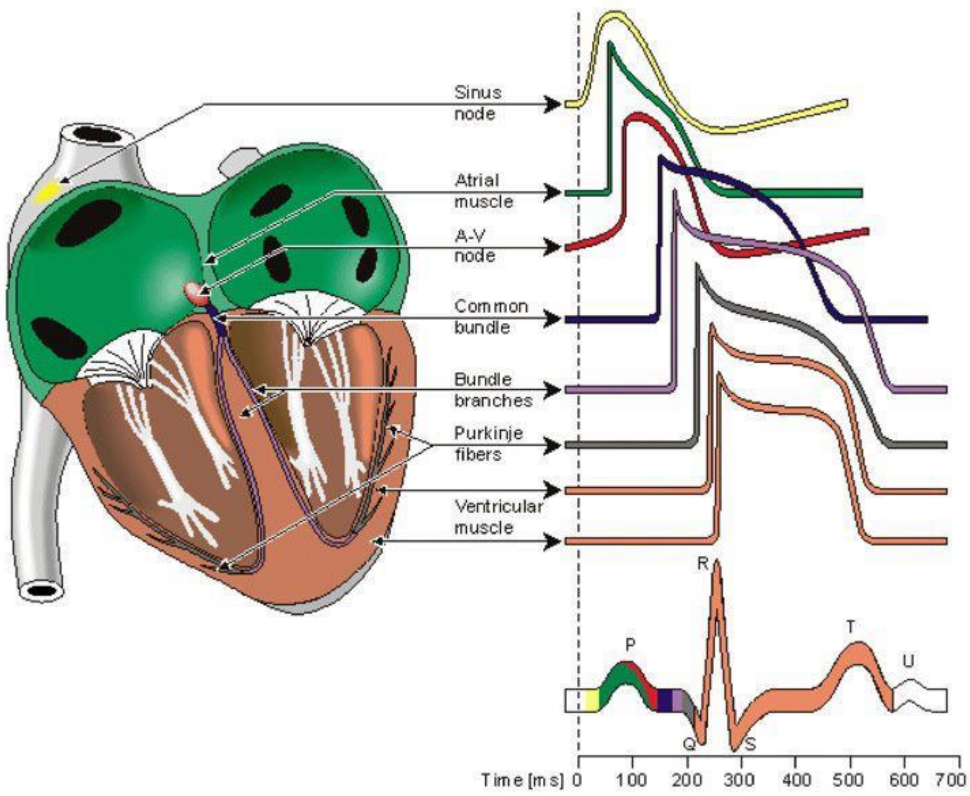



**Figure 2** ECG waveform in detail, showing different components of the ECG waveforms generated by different nodes, muscles and fibers in the heart [5]

As mentioned before, ECG is a mere representation of the electrical activity of the myocardium's cells. It can be better explained as a differential measurement across the surface of the human body, and can be thought as a measurement of the electrical potential with a vector. As can be seen in Figure 2, the typical ECG waveform contains three primary features: the P wave, the QRS complex and the T wave. The working of each part of the heart can be observed to each of these waves. The P wave represents the depolarization of the atria, thus normally being first. The PR interval is representative of the delay that happens in the A-V node giving the atria time to contract before depolarizing the ventricles. The QRS complex is the strongest wave in ECG, since it shows the depolarization of the ventricles. The T wave represents the repolarization of the ventricles. [6]

***ECG Leads:***

ECG vectors are generated by taking differential measurements of electrical potential on different locations on the body's surface. The ECG standard for clinical measurements consists of 12 leads; six of them are in the parallel plane to the body, on the chest. These leads are known as precordial ECG leads and they can be seen in the lower right corner of Figure 3, labeled V1, V2, V3, V4, V5 and V6. The remaining six leads are giving a view of the heart from the perpendicular plane. These leads are known as the frontal leads, that can be seen in the upper right corner of Figure 3, labeled as aVR, aVL, aVF, lead I, lead II and lead III. [6]

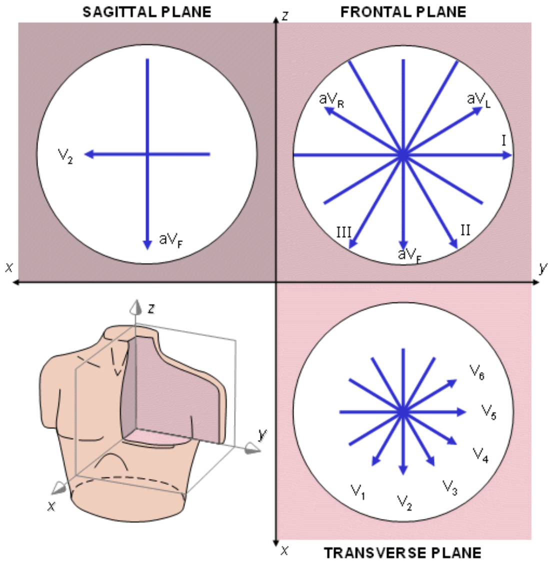



**Figure 3** Vector view of the standard 12 lead ECG. [7]

In the measurement of ECG leads, three different electrodes are placed on the body; this would be the formation of the frontal ECG leads. The electrodes are traditionally placed on the left arm (LA), right arm (RA) and left leg (LL). The 3 main electrodes form a triangle which is known as the Einthoven's triangle that can be seen in Figure 4. [6]

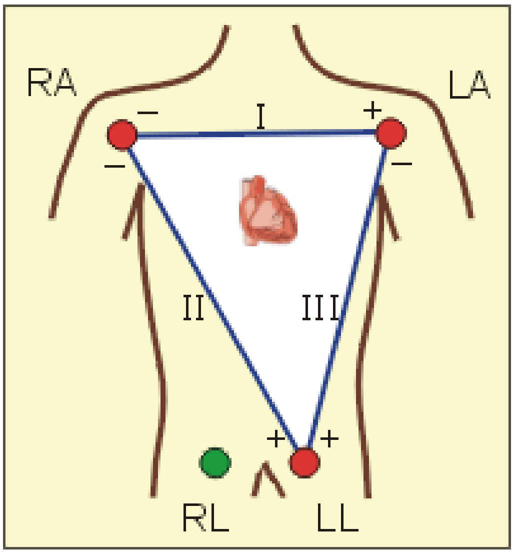



**Figure 4** Einthoven's triangle formed by the 3 main electrodes [6]

***Lead I ECG System***

Biomedical engineers have been interested in finding simplified methods for detecting chronic diseases for a long time. This led them to the research of a simple method for recognizing abnormal heart behavior. The 12-lead ECG has been used to detect heart abnormalities for many years, even though it is a bulky measurement system. As an alternative to the 12-lead system there have been studies comparing the single lead ECG with the 12 lead ECG, with quite positive results regarding the lead I system. [8]

The lead I system or single lead system provides a one dimensional low frequency signal through the application of three electrodes, two of them being active and one being a ground electrode. A possible arrangement to obtain a lead I system ECG can be seen in Figure 5. This arrangement is used by one of the most interesting and novel ECG bracelet, known as Nymi, the bracelet that uses ECG as a biometric recognition technology. [9]

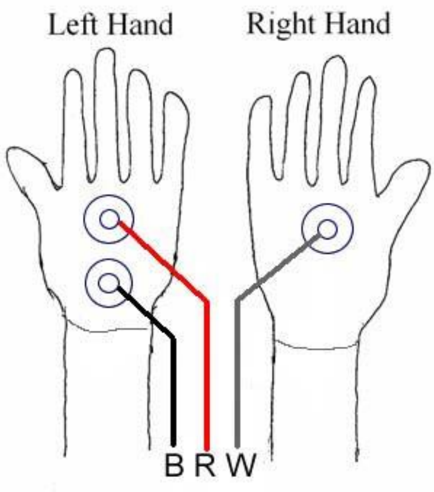



**Figure 5** Possible electrode arrangement for 1 lead ECG, where B and W are active electrodes and R is the reference/ground [9]

There exist two different types of leads in existence, the unipolar and the bipolar. The bipolar being the one that has one positive and one negative pole, while the unipolar leads also have two poles, however, as a voltage is measured the negative pole is a composite pole. The type of leads that is used in tis electrode arrangement is the bipolar leads. [10]

When the comparison was made between 1 lead ECG and 12 lead ECG, they were compared by studying a group of 2000 patients and having professionals analyze the measurements taken. Although it wasn't the intent of the study, it went on to compare the relative merits of both diagnostic methods, which can be observed in Table 1 and Table 2. [8]

**Table 1.** 12 Lead vs. Final clinical impression from 2000 patient records [8]

| Final Clinical Impression | Number | | | | Percent | | | |
|---|---|---|---|---|---|---|---|---|
| | Normal | Doubtful | Abnormal | Total | Normal | Doubtful | Abnormal | Total |
| No Heart Disease: Total | 1400 | 114 | 53 | 1567 | 89.3 | 7.3 | 3.4 | 100 |
| Possible Heart Disease: Total | 100 | 32 | 24 | 156 | 64.1 | 20.5 | 15.4 | 100 |
| Heart Disease: Total | 147 | 51 | 79 | 277 | 53.1 | 18.4 | 28.5 | 100 |

As can be seen in Table 1 the 12 lead ECG is very reliable in selecting those people free of heart disease but there is a relatively high percentage of not recognizing people with cardiovascular disease.

**Table 2.** Lead I vs. Final clinical impression from 2000 patient records [8]

| Final Clinical Impression | Number | | | | Percent | | | |
|---|---|---|---|---|---|---|---|---|
| | Normal | Doubtful | Abnormal | Total | Normal | Doubtful | Abnormal | Total |
| No Heart Disease: Total | 1259 | 144 | 164 | 1567 | 80.3 | 9.2 | 10.5 | 100 |
| Possible Heart Disease: Total | 99 | 20 | 37 | 156 | 63.2 | 12.9 | 23.9 | 100 |
| Heart Disease: Total | 140 | 53 | 84 | 277 | 50.7 | 19.1 | 30.2 | 100 |

It is apparent in Table 2 that doctors lose the ability to categorize as normal, those patients that have no cardiovascular disease. However the ability to pick up some of the more noteworthy cardiovascular diseases with this method is just as good as the 12 lead system, if we were to observe only the number it would appear that lead I is in fact slightly better. Taking this information into consideration, we proceeded to accept the lead 1 system as an acceptable method to obtain ECG measurement.

## 2.2 Skin-Electrode Interface

In order to understand the interaction between the skin and electrode, it is necessary to note that the skin is made out of three layers: the epidermis, the dermis and the subcutaneous layer. The most important interface in the interaction between the skin and the electrode is the epidermis, the outermost layer, which is divided into three layers, the most important is again the outermost layer which is called the stratum corneum, comprised by dead skin cells and is considered as a semi-permeable membrane to ions. We can observe all the different interfaces in Figure 6, as we can see there are numerous variables involved in this interface. [11]

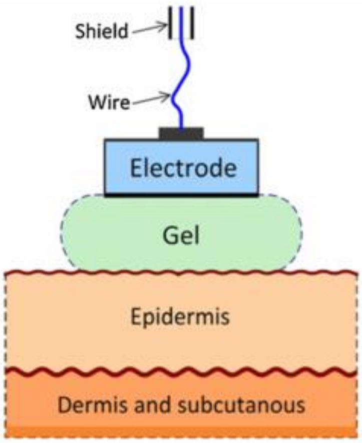



**Figure 6** Graphical model for electrode to skin interface [11]

Knowing this information, a model representing the electrical relationship between the skin and the electrode can be calculated and obtained; a representation of this model can be seen in Figure 7. In this model we can observe that the resistance of the skin and its capacitance create a parallel circuit representing the epidermis layer, more specifically the stratum corneum. [12]

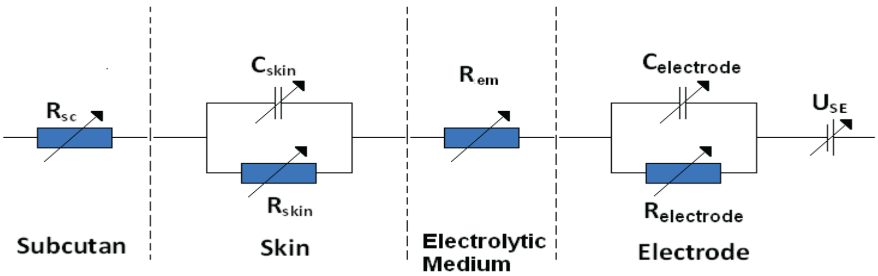



**Figure 7** Electrical model of the skin-electrode behavior [12]

The interface between the electrode and tissue is basically an ionic conductor. This conductor is the electrolytic medium that has the ability to be represented as a resistor in series. In the case of dry electrodes, the electrolytic medium will be determined by means of the presence of sweat or other forms of humidity, i.e. atmospheric humidity, etc., that may be found on the skin. In the traditional Ag/AgCl electrode there is the presence of hydrogel which reduces the resistance of the skin electrode significantly. [12]

Finally the conversion from ionic current to electric current takes place, where the electrode and the electrolytic medium make contact, caused by means of an electrochemical reaction that creates an ion-electron exchange. This ion-electron exchange has the ability to be seen as a capacitor with a double layer of charges created at the interface. [12]

The electrode polarization occurs by means of a displacement of positive and negative charges on the interface of the skin and electrode to opposite sides of the electrode. The impedance has the ability to be obtained by means of using current that flows via the tissue, and the voltage drop being directly influenced by the electrode polarization. [12]

## 2.3 Electrodes

### 2.3.1 Wet Electrodes

Ions are used as charge carriers in the body; thus picking up bioelectric phenomena on the occasion of the ionic charge carriers convert ionic current into electric current. This transducing function is facilitated on the occasion of using electrodes that consist of conductors with liquid ionic solutions in the body. [13]

The interaction between a metal and a solution with its ions produces a change in the ion concentration. This causes a difference in electrical potential between the electrolyte near the metal surface and the rest of the solution. This potential difference is known as the half-cell potential, established between the majority of the electrolyte and the metal. It has been observed that different types of materials have different characteristics as can be seen in Table 3. [13]

**Table 3**. Half-cell potentials for materials and reactions encountered in biopotential measurement [13]

| Metal and Reaction | Half-cell Potential, V |
|---|---|
| $Al \rightarrow Al^{3+} + 3e^{-}$ | –1.706 |
| $Ni \rightarrow Ni^{2+} + 2e^{-}$ | –0.230 |
| $H_2 \rightarrow 2H^{+} + 2e^{-}$ | 0.000 (by definition) |
| $Ag + Cl^{-} \rightarrow AgCl + e^{-}$ | +0.223 |
| $Ag \rightarrow Ag^{+} + e^{-}$ | +0.799 |
| $Au \rightarrow Au^{+} + e^{-}$ | +1.680 |

Silver-silver chloride (Ag/AgCl) electrodes have similar characteristics to a perfectly non-polarizable electrode, and are very convenient for biomedical applications. This electrode consists of a silver base covered by means of a layer of ionic compound silver chloride. Due to the fact that silver chloride is relatively insoluble in liquids, the surface remains stable. Since there is minimal polarization on Ag/AgCl electrodes, this helps to avoid smaller noise caused by means of low frequency along with the fact that motion artefacts are greatly reduced compared to other polarizable electrodes, making this electrode the standard ECG electrode. [13]

It is imperative to know that the perfectly polarizable electrodes are those electrodes in which there is no charge crossing the electrode-electrolyte interface when a current is applied. The electrode behaves similarly to a capacitor, where the current across the interface is a displacement current. The perfectly nonpolarizable electrodes are the electrodes in which current may pass freely across the electrode electrolyte interface, making the transitions without any need for energy. These electrodes behave more similarly to a resistor. [14]

The electrical representation of this type of biopotential electrode is usually nonlinear and to be characterized in linear models it is required that they operate at low currents and potentials. Under ideal conditions the Ag/AgCl electrode can be represented in an equivalent circuit as can be seen in Figure 8. [15]

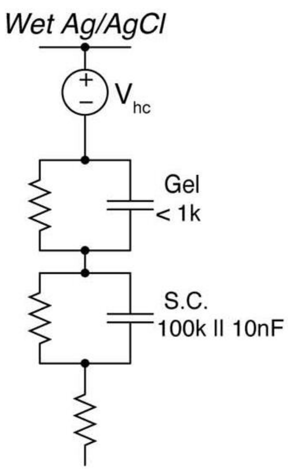



**Figure 8** Electrical model of skin-dry electrode model, where $V_{hc}$ is the half-cell potential, and gel and skin are modeled in parallel capacitive and resistive elements. [15]

### 2.3.2 Dry Electrodes

The dry electrode's usefulness and performance may be divided into two categories. The first depending in regards to the interface between the electrode and the skin, the electrode should be evaluated in regards to its comfort along with its utility. The second is relating to the signal quality, based on the amount of noise. This technology requires mechanical proposals in order to be secured on the patient. For this reason dry electrodes have not substituted the standard ECG electrode, which is the wet Ag/AgCl electrode. [15]

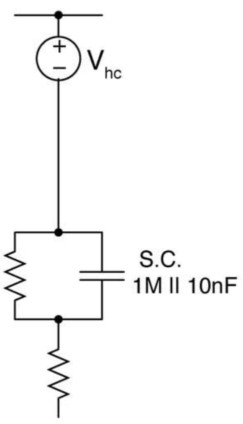



**Figure 9** Electrical model of skin-dry electrode model [15]

The dry electrode needs to be used along with an actively shielded amplifier, which has to be chosen for its immunity against outside interference along with line noise. [16] The electrical model for the purpose of the dry electrode-skin interface can be seen in Figure 9. Due to this electrical model and the noise that exists in it, it is recommended to use the topology shown in Figure 10, which conforms to many of the published amplifier circuit for this type of electrodes. [17] [18] [19]

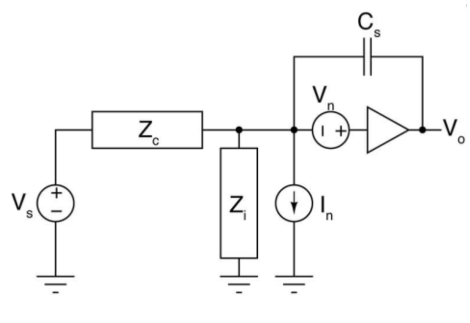



**Figure 10** Dry electrode amplifier circuit noise model [15]

The main difficulty in this type of sensor is the motion artifacts, especially in mobile, wearable ECG sensor systems. This is due to the fact that this type of artifact is subject to human variability and it is ill-defined, unlike expected parameter along the lines of noise, gain and power consumption. These other artifacts have the ability to be easily simulated and readily measured, thus being possible to design a circuit minimizing these artifacts. In addition, different types of dry electrodes are subject to different types of interference and noise. The lack of measurable facts along with the difficulty to measure them resulted in less devotion in this area. [15]

***Textile Electrode***

Textile materials are usually insulators; however, on the occasion that conductive yarn is attached to the fabric during their manufacture, they become conductive. These electrodes do not require the use of gel in order to create a connection with the skin, since they behave similarly to dry electrodes. The conductive textile has the ability to be made by means of silver coating yarn or filaments that have the ability to be embroidered into the textile. The majority of the textiles that are used are polyester and polyamide, due to the fact that they endure abrasion, dry fast and absorb little amounts of moisture [20].

Other benefits of this type of electrodes include the fact that they are quite recommended for the purpose of long term use since they do not irritate the skin, they are lightweight and easily washable. However it is important to acknowledge the high impedance in the electrode-skin interface, from 1 $\Omega/cm^2$ to 5 $\Omega/cm^2$. In order to have a reference point the more commonly used Ag/AgCl has an impedance of 3.5 $\Omega/cm^2$. [3] [21] [22]

Considering that textile electrodes lack electrolytic properties, the electrode polarization impedance increases because of dry or greasy skin surfaces, which would cause more perceptible changes in the measurements, but in previous studies developed by Seoane and Välimäki it has been proven that there have been no remarkable differences observed in the resulting signals. [23] However, during this study the electrodes were humidified, by adding water,to improve the electrical interface between the electrode and the body. By humidifying the electrode the electrode polarization impedance was not much larger than that of the Ag/AgCl electrode. [23]

Remarkably, further studies developed by Cömert, Honkala et al have shown that the signal quality under no motion conditions of dry textile electrodes, are comparable to other commercial electrodes. This shows that the main problem for the textile electrodes is the motion artifact, due to the fact that they are not attached via glue to the skin. Placing the electrodes in carefully selected anatomical locations could eliminate or minimize the effects of motion to a certain degree. [24]

A recent study by Cömert et al. devised a method to quantify motion artifacts on ECG measurements, and by utilizing this quantification method there were further studies performed to observe the effects of using padding along with textile electrodes to decrease these artifacts. These studies lead to the conclusion that by utilizing padding it is possible to reduce the motion artifact noticeably. The pad is recommended to be put between the band and the textile electrode, and the recommendation is to use a soft filling. Regardless of the lining, the study also concluded that the ideal pressure exerted by the attachment method was between 15 to 20mmHg, in order to stabilize the electrode and obtain the best signal quality possible. Coincidently this amount of pressure falls into the “tight but comfortable” range for wearable products. [25]

## 2.4 Signal Processing

### 2.4.1 Analogue Signal Processing

Analogue signal processing is oriented for signals that have yet to be digitized. This type of processing involves both linear and nonlinear electronic circuits. Examples for linear circuits are passive filters, active filters, integrators and delay lines, and for nonlinear circuits there are compandors, voltage controlled filters and phase locked loops [26]

***Low Pass Filter***

The main objective of a low pass filter is to filter all the high frequency signals and attenuate them. The cutoff frequency is the boundary in which a system starts to reduce or attenuate a signal. Low pass filters can be applied to signals both digitally and with hardware, the most common electronic low pass filter is a resistor-capacitor circuit (RC circuit). The RC circuit is composed, as the name says, of a resistor and a capacitor as can be seen in Figure 11. To set the cut off frequency it is imperative to calculate the appropriate value for the resistor R and the capacitor C. [27]

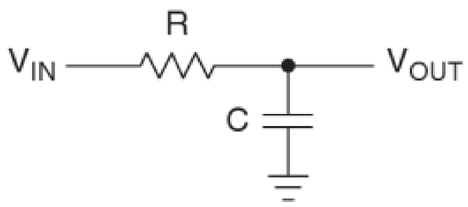



**Figure 11** Circuit schematics for a first order low pass RC filter,where there is only one pair of resistor and capacitor between input (Vin) and output (Vout) [27]

Values for the components are calculated according to equation (1.1), where fc represents the cut-off frequency of the filter.

$$f_c = \frac{1}{2\pi RC} \tag{1.1}$$

Analog filters are recommended in ECG circuits since we are measuring data that is time variant whose output and input are both analog. However the most important reason is that real time filtering is preferred and recommended in ECG measuring, due to the necessity of precision and fidelity to the source data. Along with this recommendation, it is also recommended to use digital filtering fine tuning method, since they have higher accuracy and no drift due to component variations. For this type of filtering it is important to remember that high performance ADC is required and due to the computation time the output will be delayed. [28]

***Integrators***

Integrators are constructed to integrate signals with respect to time. This is a form of first order low pass filtering, which has the possibilities to be performed both in the continuous time domain and the discrete time domain; however, in the latter it would be approximated. An integrator produces a steadily changing output voltage for a constant input voltage; the model circuit can be seen in Figure 12. [29]

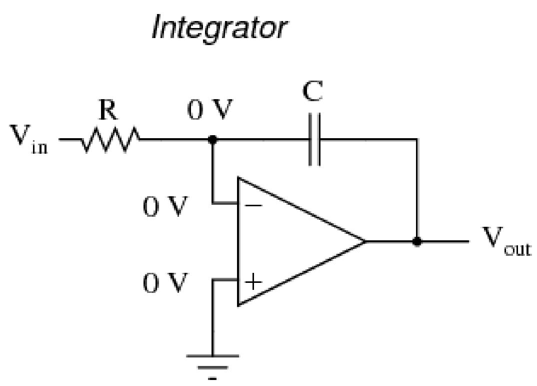



**Figure 12** Integrator circuit. [29]

Values for the components are calculated according to (1.2), where $v_{out}$represents the voltage output of the operational amplifier, $v_{in}$ represents the input voltage, t would represent time. R represents the resistor and C stands for the capacitor, though it will also give the output voltage at t=0. [29]

$$\frac{dv_{out}}{dt} = -\frac{v_{in}}{RC} \tag{1.2}$$

### 2.4.2 Digital Signal Processing

This processing is for digitized discrete time signals, and is done by computers or digital circuits, such as ASICS or FPGA's. The main objective of digital processing is to measure, filter and compress analogue signals. The first step is usually to convert the analogue signal to digital using an ADC. This process is usually more complex than analogue processing, however, the application of computational power to digital signal processing has a number of benefits, such as error detection and correction in transmission. [30]

***Wavelet Denoising***

Wavelet is an oscillation that behaves like a wave with amplitude that begins with zero, increases and then decreases back to zero. Wavelets are purposefully manufactured to have specific properties to make them useful for signal processing. [31]

The wavelet transform accomplishes a correlation analysis; for which it is expected the output to be maximal when the input signal is the most similar to the mother wavelet. Lessening the wavelet transform will eliminate the low amplitude noise or unwanted signal in the wavelet domain, and then an inverse wavelet transform will retrieve the desired signal with little loss of detail. [31]

There exists two types of thresholds; soft and hard thresholds. It is known that soft thresholding provides for a smoother result compared to hard thresholding. Meanwhile hard threshold provides for a better edge preservation. Sometimes it is recommended to apply the soft threshold to some details and the hard threshold to the remainder. [31]

### 2.4.3 Biopac MP36

The Biopac MP36 is a data acquisition hardware that has built in amplifiers, made by BIOPAC systems. The objective to record and condition electrical signals from the human body, be it heart, muscle, nerves, respiratory system, etc. The data that is usually measured are extremely small, sometimes in the microvolt range. This hardware amplifies these signals, filters unwanted signals and noise, to later convert the data into a set of numbers that the computer can read. Finally the Biopac software displays the information as waveforms. [32]


**Figure 13** Biopac MP36 System, enabling measurement and display of biosignals. [33]

The Biopac student lab system is aimed to help understand and explain the fact that electricity is constantly flowing throughout the living body to students. [32]

# 3 METHODS AND MANUFACTURING

## 3.1 Electrical Design

It is imperative to make an appropriate plan before proceeding with the design of the circuit. The main concern at the beginning was to decide how we would handle the signal that was recouped from the person. We have established in the previous chapter that the lead I system can be used for diagnostic purposes, with a signal quality quite comparable to that of the 12 lead system. By utilizing the lead I system, we know that we will need to work with three inputs to the electronic circuit, which still needs to be conditioned and processed to obtain a visible ECG.

Taking into consideration that comfort is highly recommended and desired in this project, it would be ideal to make the size of the bracelet and circuitry as small as possible. Also it should preferably allow further developments, have the possibility for long recordings and have a user friendly interface. After much consideration, it was decided to use existing technology, such as cellular phones and computers, to obtain and process the signal. In these systems there can be an endless array of existing programs and some that can be developed for analyzing the ECG waveform.

Having a clear view of where we want to process the information and how it will be obtained, we proceeded to consider how the communication system would be designed. In the beginning it was thought to use a dovetail joint as can be seen in Figure 14, to make a proper connection and also avoiding motion artifacts in the electrode connection. However after prolonged consideration of the advantages and disadvantages, we came to the conclusion that it would not be required or helpful to use this type of joint. The main reason was that it would have to be done with less than optimum material, aluminum, and it would also be too restrictive for the user, giving an unnecessary imprisoned feeling.

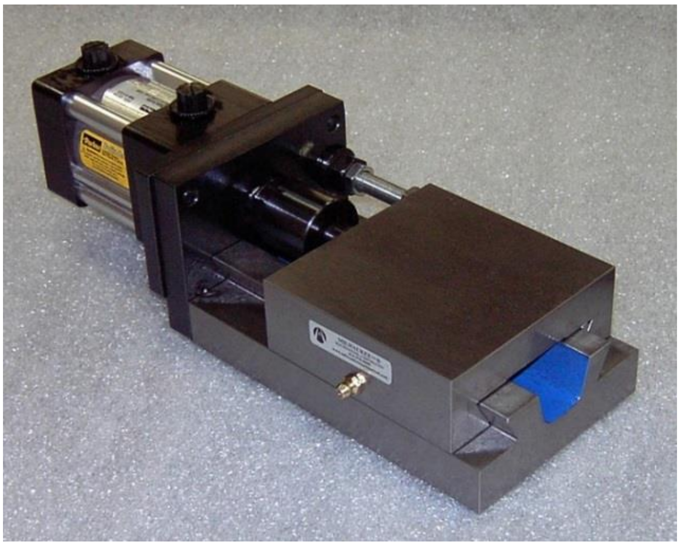


**Figure 14** Dovetail joint, which was intended to be used as a fixture for electrodes

After considering various possible communication systems and connections, it was decided to follow the advice of a project to utilize a headphone jack connection for the data transfer to the data recorder. [34] When utilizing the headphone jack it would be important to amplify the signal sufficiently, along with filtering the signal before we transfer the information on the device, which in this case would be a portable computer. The amplification and filtering can be seen implemented in Figure 15, in which the signal has a low pass filter to allow signals under 40 Hertz to pass through to the output, while taking away most of the electrical noise. This value was selected for the low pass filter due to the fact that ECG frequency is usually less than 40 Hertz. [34] The operational amplifiers selected to be used for this project were the LM324N, given their four op amps in one package, their large DC voltage gain, wide power supply and the fact that the power drain is suitable for battery operation [35].

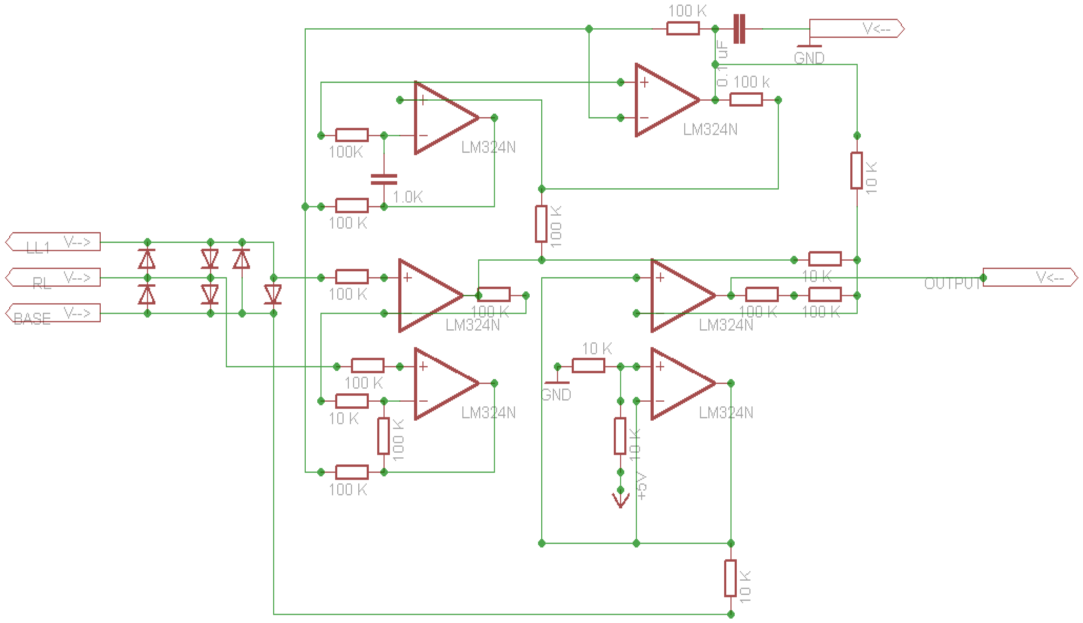



**Figure 15** Electronic schematic for ECG amplifier. Circuit schematics includes filtering, amplification and redundant protection for the patient [36]

The circuit has redundant protection for the patient, as can be seen by the diodes added to the inputs and by the 100 KΩ resistors before the connection to the body. These resistors are there in case there is a connection to a bad node, the current would be low. Most of the amplifiers used in the circuit are for noticeably amplifying the biosignal in this circuit it is approximately 1000. However, some are used as a filter, with a CMRR of 80 dB and also as a buffer to provide a constant voltage of half the power supply providing a reference voltage.

At this time that the communication system between the electronic circuit and the data recorder has been decided, there was a need to find a proper medium to transfer data between the electronic circuit and the bracelet. This system has to be ergonomic, highly conductive and have good data transmission. In order to obtain all the previous characteristics we have considered various manners to transmit the data, from Bluetooth to using wires. Finally it was concluded that a contact transmission with a hook and loop fastener would be ideal and from the circuit we transmit to the computer using the microphone jack.

Taking the different types of communication and the potential risk of receiving a shock through the electrodes into consideration, it was decided to also implement a system of diodes in the circuit, making it into a simplex communication, meaning that the data will only be allowed to be transferred in one direction, as can be seen in Figure 16. The schematic was designed in such a way that it can be easily miniaturized and added to the bracelet without making it bulkier, or uncomfortable. Finally the circuit has also been intended to be easy and cheap to manufacture, so as to make this type of technology readily available for anyone.

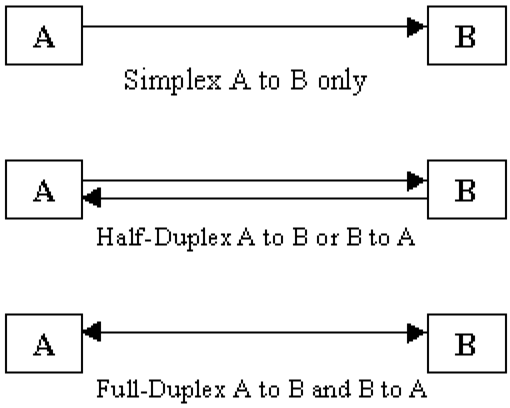



**Figure 16** Basic communication modes of operation [37]

The decision to utilize the headphone jack created the possibility of recording the ECG waveform via any device that supports microphones and a recording program. This enables the use of audio recorders, cellular phones, tablets and computers to capture the data that we can later process by using digital audio editing shareware. Finally using the same software to record the ECG can be used to observe and analyze it, this could also be done in real time.

## 3.2 Mechanical Design

### 3.2.1 Electrode Material Selection

After obtaining the data it is needed to decide which type of electrode would be best to make the measurements. As mentioned before the Ag/AgCl electrodes contain an electrolytic medium with conductive properties in order to reduce the polarization impedance's value; in dry-textile electrodes the lack of this electrolytic medium increases the impedance, which in turn contributes to capacitive leakage.

In order to observe the behavior of textile electrodes and their use in bio-measurements various studies applied to both types of electrodes were investigated. One of the first studies, by J.C.M Ruiz, found was a comparison between the performances of the Adistar knitted textile electrode, the impedimed electrode and the red dot 3M repositionable monitoring electrode, with respect of the recorded impedance spectra.

As can be seen in Figure 17 there is a noticeable difference between the electrodes, especially as we approach larger frequencies, but in the end result we can consider that the textile electrodes are usable for the purpose of our applications due to their good reactance in all frequencies and relatively stable resistance over the frequency range of ECG systems.

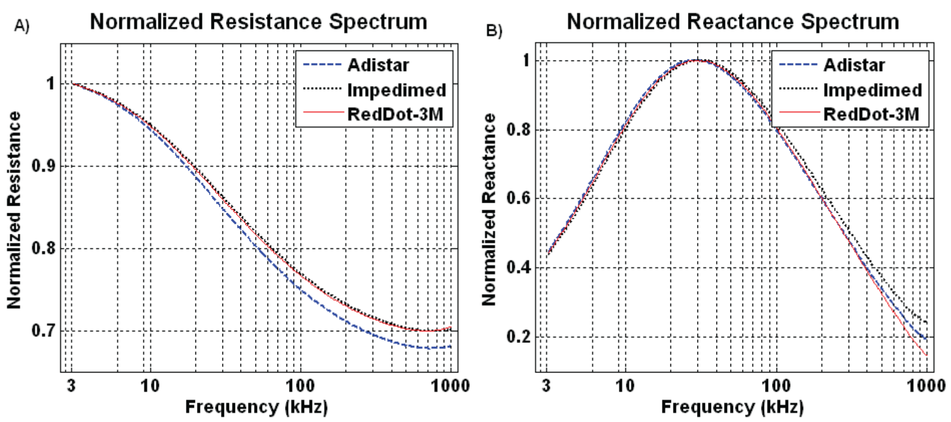



**Figure 17** Comparison of resistance and reactance spectrums of the Adistar, Impedimed and RedDot-3M electrodes [3], Difference between the electrodes become more noticeable in higher frequencies.

In 1940 a mathematical equation was developed that fits the obtained electrical bioimpedance measurements. This equation is commonly used to not only represent but also analyze the electrical bioimpedance data. This equation can be observed in the equation 1.8, with $R_\infty$ as the ideal resistor, $R_0$ as the static resistor, ω as the angular frequency, α is an exponent parameter and τ as the time constant, to give us the complex value containing resistance and reactance $Z_{cole}$. [38]

$$Z_{cole}(\omega) = R_\infty + \frac{R_0 - R_\infty}{1 + (j\omega\tau)^\alpha} \tag{1.3}$$

The Cole formula (1.3) gives us some knowledge of what is taken into consideration for this equation before we proceed to evaluate the mean values for the Cole parameter, which has been estimated from each type of measurement and from the subject used in a study done by KTH technology and health along with the University of Boras. [3]

**Table 4** Mean values of the estimated Cole parameters from EBI measurements, where $R_0$ is the static resistor, $R_\infty$ ideal resistor constant, $f_c$ is the cut-off frequency and α is an exponent parameter. [23]

| | $R_0$. | | | | α | | | | $f_c$ | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| **S** | 3M | Tex | 3M_G | Tex_G | 3M | Tex | 3M_G | Tex_G | 3M | Tex | 3M_G | Tex_G |
| **1** | 424.0 | 445.3 | 447.9 | 439.4 | 0.728 | 0.711 | 0.714 | 0.713 | 31.1 | 29.8 | 29.6 | 31.1 |
| **2** | 461.4 | 439.5 | 478.4 | 464.6 | 0.709 | 0.717 | 0.702 | 0.706 | 35.7 | 37.1 | 32.7 | 36.9 |
| **3** | 581.9 | 567.3 | 595.3 | 581.4 | 0.714 | 0.721 | 0.701 | 0.721 | 41.7 | 42.0 | 40.5 | 40.5 |
| **4** | 449.4 | 451.6 | 465.5 | 463.7 | 0.705 | 0.713 | 0.713 | 0.702 | 32.9 | 33.0 | 33.4 | 33.4 |

The values in Table 4 clearly indicate that the mean of the differences are less than 5%, with the exception of $R_0$. However in α we observe values lower than 2.4%, unlike $f_c$ whose deviations are closer to 4.4%, due to the fact that the textile electrode creates an overestimation of this value.

After concluding the viability of textile electrodes we proceed to analyze the findings of a published experiment, by Puurtinen, Komulainen and Kauppinen, in regards to the noise level from biopotential measurements, in which the signals were processed in Matlab. The parameters measured and compared were the power spectral density (PSD), which helps to observe the behavior of the spectral components of the signal and the root mean square (RMS) noise.

There are three main possibilities of utilizing textile electrodes, the first is having the electrode dry, then with a moist electrode and finally with covering the electrode with hydrogel. In Figure 18 we can observe the RMS noise in respect to the electrode size, with all of the 3 types of electrodes. From this figure we can conclude that the RMS noise value for the purpose of wet textile electrodes and with hydrogel is quite low compared to the dry version. Also we observe that the RMS noise level increases as the electrode size decreases.

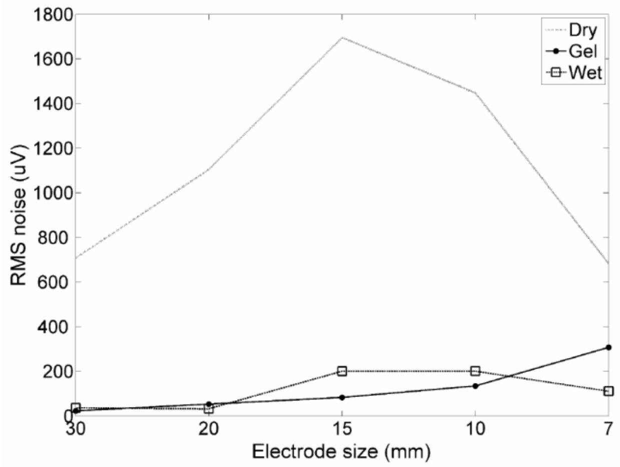



**Figure 18** RMS noise as a function of electrode size for dry, wet textile electrodes and textile electrodes with hydrogel. This figure indicates that RMS noise levels are quite low for both wet textile electrodes and textile electrodes with hydrogel for different electrode sizes, whereas dry electrodes have significantly higher noise for all electrode sizes [39]

We proceed to show the figures in which we have the ability to observe the power spectral density (PSD) of signals recorded with dry textile electrode (Figure 19), wet textile electrode (Figure 20) and textile electrodes with hydrogel (Figure 21). In these figures the size of the electrodes are relevant and made a noticeable difference, specifically with wet textile electrodes. According to the figures below it can be said that the bigger the electrode size, the bigger is the PSD value.

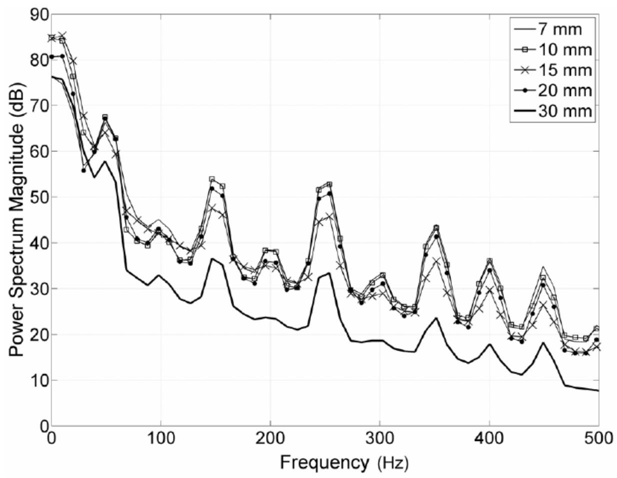



**Figure 19** PSD of biopotential signals recorded with dry textile electrodes of different sizes of 7mm, 10mm, 15 mm, 20 mm and 30 mm [23]

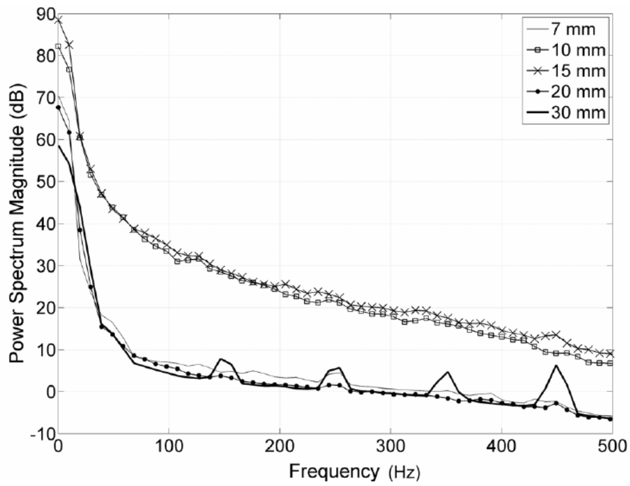



**Figure 20** PSD of biopotential signals recorded with wet textile electrodes of different sizes of 7mm, 10mm, 15 mm, 20 mm and 30 mm [23]

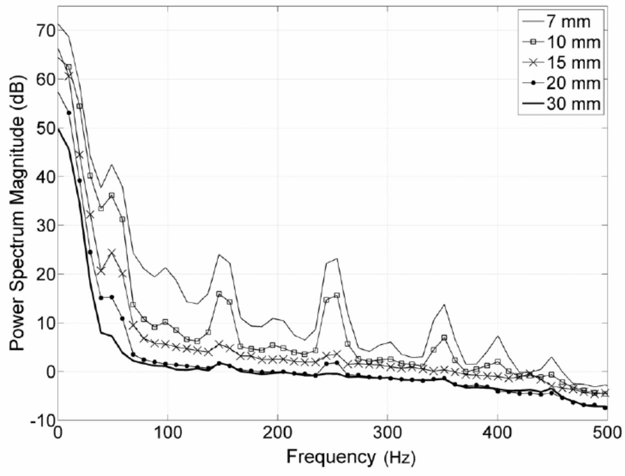



**Figure 21** PSD of biopotential signals recorded with textile electrodes covered with hydrogel of different sizes of 7mm, 10mm, 15 mm, 20 mm and 30 mm. [23]

As can be seen from Figure 19, in dry electrodes, the sizes of the electrodes do not noticeably affect the measurements; however, the largest electrode does have lower noise. While in the wet textile electrodes (Figure 20) the 10 and 15 mm electrodes have the same noise level, the same can be said of the 20 and 30 mm electrodes whereas the 7 mm is the electrode size with the least amount of noise. In the final analysis the electrodes with hydrogel (Figure 21) have the behavior of an increase in noise as the size decreases, being the most consistent electrodes with the changes in size.

### 3.2.2 Bracelet Design

***Design Requirements***

In recent years there have been numerous attempts and proposed ECG bracelets, all being successful in realizing their goals. However, most of them differ from the one proposed in this thesis when it comes to comfort, some differ in the final objective for the device. Before designing the proposed ECG bracelet it was decided to study the three most successful ECG bracelets; to learn from their advantages and disadvantages, thus being able to create a working and more comfortable hybrid ECG bracelet, while complying with the results of previous designs.

One of the textile electrodes used for measuring ECG that gave the best results was a randomly embroidered electrode made with a sewing machine, done by Pola and Vanhala. The suspected reason was that the conductive yarn was on top of the electrode, hence providing a better contact with the skin by being on relief, (Figure 22). The conclusion of the study was that ECG could be measured by using textile electrodes, mentioning specifically the "very good" results obtained by this electrode. Though it makes the emphasis on including a preamplifier to the electrode and that it is necessary to investigate on a manner to improve contact between the electrode and skin. [40]

**Figure 22** Different types of electrodes used in ECG measurement. The first three electrodes were made industrially but the fourth electrode is the handmade embroidered conductive yarn [40]

After the Pola and Vanhala study, there has been a study, by Cömert, Honkala and Hyttinen, regarding the effects of both pressure and padding for ECG measurements with textile electrodes. In the literature, various degrees of pressure between 5 and 25 mmHg was tested, along 5 different types of padding of differing heights and grading were investigated. The study concluded that the use of padding greatly improved the measurements' data signals, by reducing considerably motion artifacts, with not much difference between the types of padding used. The recommended pressure for measuring ECG is from 15 to 20 mmHg. [25]

Continuing the analysis of different types of electrode bracelets, there is a different type of electrode bracelet that gave rather successful results while measuring without any gel or moisture, as can be seen in Figure 23. The electrode was made out of a 3 mm thick polyethylene tereftalate (PET) polymer with a thin conductive layer of titanium nitride (TiN). However, even if the signal was similar to that of the Ag/AgCl electrodes, the main problem with this type of electrode was the electrochemical reaction with saline solutions. When the electrode was submerged in a solution simulating sweat interaction on the skin for at least 9 days, the electrode corroded. [41]

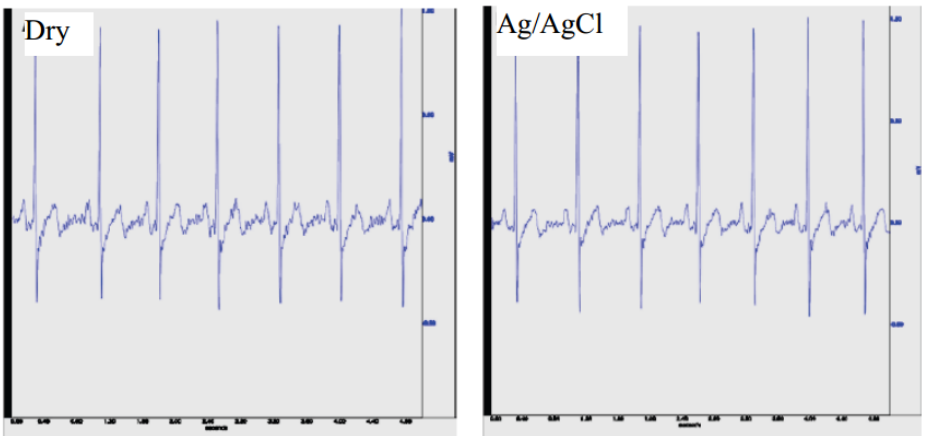



**Figure 23** Comparison of PET bracelet electrodes and Ag/AgCl electrodes [41]

Silver is known for not being reactive, specifically to oxygen or water, making it an ideal metal to be used for the conductive textile. Silver has many excellent characteristics; it is highly conductive, dissipates static, anti-bacterial, anti-fungal, has a really high ductility and is also hypoallergenic if using 99% pure silver, also known as fine silver. Given all the characteristics mentioned here and considering the real life scenarios of a biomedical device, it was decided that it would be ideal to utilize fine silver.

The need for biomedical devices to be long lasting, easy to clean, hypoallergenic and made of the highest quality materials cannot be stressed enough. There have been numerous standards in place regarding biomedical devices mainly by ISO 13485, ISO 14971 and ISO 13485:303. [42] Even when this specific biomedical device would be classified as a class I due to its low risk and noninvasive nature, it is necessary to consider all the possible repercussions that may be caused by using less than adequate materials. A few examples could be a possible tetanus infection due to an abrasion with an oxidized metal, or an allergic reaction to the materials of the bracelet.

Nymi utilizes the lead I system to obtain the ECG waveform, as can be seen in Figure 24, and later uses it for authentication. In the whitepaper it is clarified that the main reason for Nymi to not be used as a medical device is because the hardware does not fall into medically approved ECG technology. However they do plan to expand to the medical sector with future generations of Nymi, due mainly to its high quality data acquisition. [43]

**Figure 24** Nymi bracelet authentication system. [43]

Though the objective of the Nymi electrode is for public and general use, it would be interesting and more beneficial to apply this high quality product into the medical sector. It is important to also remark that in the Nymi white paper there is mentioned that there is no alternative for continuous ECG monitoring and its current inconvenient measuring method with numerous electrodes. These points are valid and they also present an inconvenience in the design of this bracelet. Mainly because of overall comfort it was decided to have one bracelet opposed to two and the objective of the proposed bracelet would be for periodic or occasional ambulatory measurements.

There is an often overlooked problem that can arise from ECG bracelet electrodes; there exists a possibility of promoting carpal tunnel syndrome. The syndrome can be triggered by the fact that the bracelet creates a centralized pressure point on top of the carpal tunnel. There are numerous ways of reducing the possibility of causing the syndrome, the most common methods are either using a splint or by using padding. A splint is quite uncomfortable for daily use, however keeping in mind that padding does improve the quality of the signal acquired, padding would be the ideal solution. By proceeding to make the padding thicker, the risk of carpal tunnel syndrome could be reduced noticeably, while improving the data acquisition.

After analyzing all the different advantages, disadvantages, and the recommendations from the three electrodes, it was proceeded to start the design and establish the limitations of the proposed ECG bracelet. Though not all the recommendations where heeded, most of them found a place in our bracelet. The main design objectives are summarized in Table 5.

**Table 5** Bracelet's characteristics comparison

| Characteristics | Embroided textile electrode | TiN electrode | Nymi bracelet | Proposed ECG bracelet |
|---|---|---|---|---|
| Incorporated amplifier circuit | | | ✓ | ✓ |
| Padding | ✓ | | | ✓ |
| Withstands sweat | ✓ | | ✓ | ✓ |
| Hypoallergenic | ✓ | | ✓ | ✓ |
| Good signal quality | ✓ | ✓ | ✓ | ✓ |
| Continuous measurement | | ✓ | | |
| Convenient measurement method | | ✓ | | |
| Median nerve friendly | | | | ✓ |

***Bracelet Modeling***

For the design we had to make a preliminary design for which CATIA, a computer assisted design and manufacturing program, was utilized to create a computer aided model. While designing the model the previous characteristics were not the only design characteristics taken into consideration, but we also took into consideration the ability to increase the pressure of the bracelet and ease of adjustment by utilizing a fabric hook and loop fastener as a securing device. Considering it would be made flexible, it would not cause problems when adjusting to any wrist size.

As can be seen in Figure 25, the bracelet was designed to be relatively thin, measuring 3 centimeter of width while having 3 millimeter of thickness and 27 centimeters of length. The padded part, which can be seen in the figure in gray coloring, would be the place where the silver textile will be situated, while the long black part would be the fabric hook and loop fastener. As can be seen the pad/electrode is longer than needed so that later the amplifier circuit can be integrated to the bracelet.

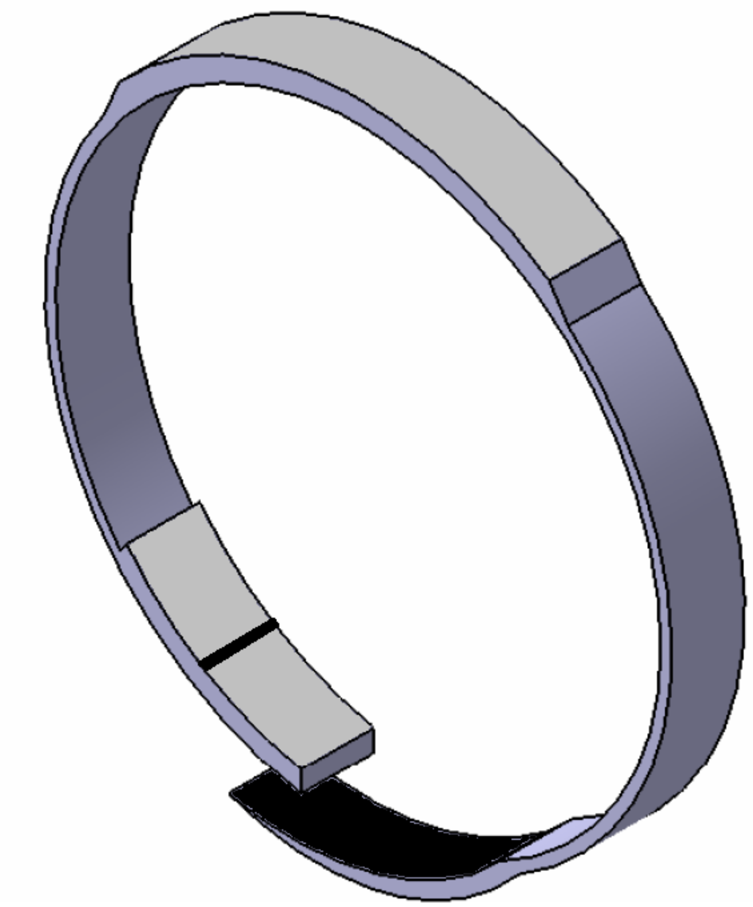


**Figure 25** Computer design of proposed ECG bracelet.

Having a design and model for the manufacture of the bracelet is essential; however it is also required to know the materials that will be used for the manufacture of said bracelet. The main body of the bracelet is recommended to be made out of polyester and polyamide, mainly due to the fact of ease in cleaning and comfort. Finally, the padding that was decided was the foam called Pudgee, which works similarly to a memory foam.

The most difficult decision came when deciding the materials that will be used for the transference of data, since the bracelet will have 3 places to interact with human skin, one for each electrode. After doing extensive research on the conductive materials available and testing them, it was decided to utilize either of the conductive textiles named Technik-tex P130+B or the Super-tex P-180 +AT, both having good conductivity, they are elastic and quite comfortable to wear for prolonged periods of time. [44] [45]

The next part in the design was the communication system between the bracelet and the recording device. For this we found an electrically conductive hook and loop coated entirely of silver, which would be basically an electrically conductive hook and loop fastener. This conduit would be separated so as to allow 3 signals through the same space. The interconnections of the bracelet will be done using conductive fabric tape, which will go through the spine of the bracelet. [46] [21]

In Figure 26 we can observe that the underside of the bracelet has the electrically conductive loop part of the fastener. This conductive hook and loop fastener is divided into three sectors, each one for every lead needed, the division in real life will be slightly thicker to avoid stray loops causing noise.


**Figure 26** Schematic for the electrode placement of the ECG bracelet

In Figure 26, we can also observe the relationship of each sector and to which lead it corresponds, which allows for a simpler interface for when we connect the bracelet to the amplifying circuit. Remembering Figure 5 we can relate number 1 and 2 to the active electrodes B and W respectively, leaving 3 to be the ground. It seems necessary to clarify that when the circuit is integrated, only 2 outputs will be needed, the signal transferring channel and the ground.

## 3.3 Electrical Simulation

Before proceeding to manufacturing the bracelet, of both the mechanical parts and the circuitry, it is imperative to simulate the schematic and test the materials that will be employed in this product. For this we will test the parts separately and after manufacturing the prototype we will realize the final testing, which shall be reported in results.

The first test that was realized was the simulation of the designed circuit, since we ran into some problems with the simulating the circuit in freeware version of Eagle, we proceeded to utilize Livewire. The act of importing the schematic from Eagle was quite straightforward and simple, thus avoiding any errors when copying the circuit. Once the circuit was ready, the simulation was run with different function generators, though we weren't able to simulate the exact ECG waveform, we did replicate the frequency and the amplitude. As can be seen in Figure 27, the simulation was successful, which gave us the information needed to proceed to the manufacturing stage.

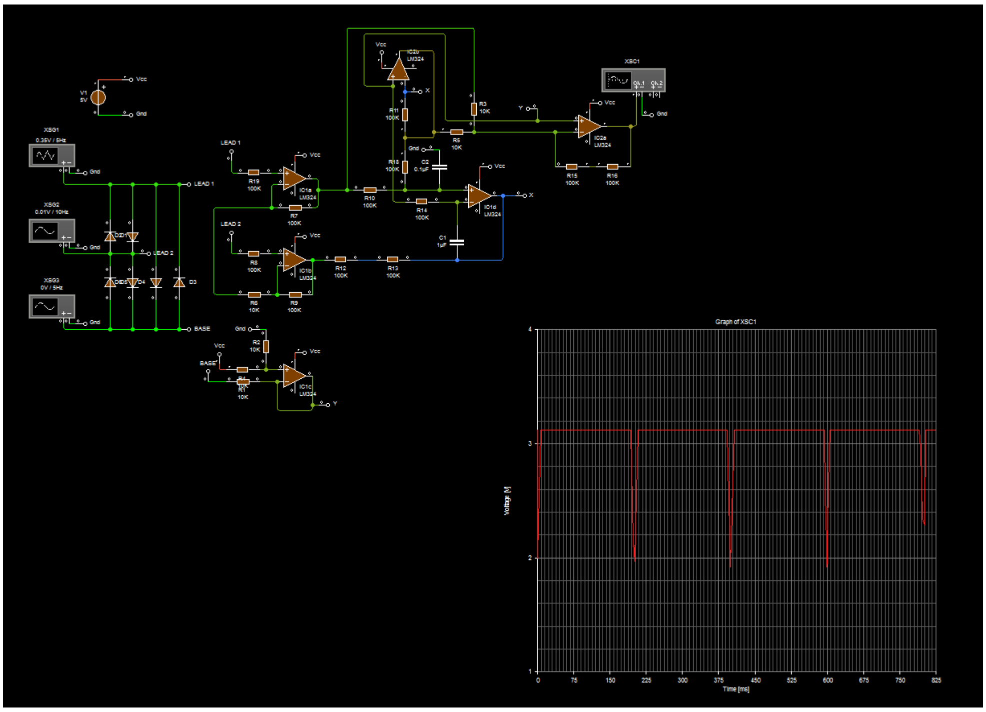


**Figure 27** Simulation results of ECG circuit

## 3.4 Manufacturing

The manufacturing process is greatly aided when there is a good design and proper planning before starting. The fact that there was no need for changes during the testing and the decisions taken during this process have simplified the work that will take place during the manufacturing.

We proceeded to work on the circuit since it was the part that we considered to be the most laborious one. As originally thought it took quite some time to get all the components in place and soldered, and in the first couple of tries there were some problems in the functionality. The two times this happened it was both due to the fact that there was a problem while soldering and a connection was not done. As can be seen in Figure 28, the circuit was successfully done and was ready to start measuring.

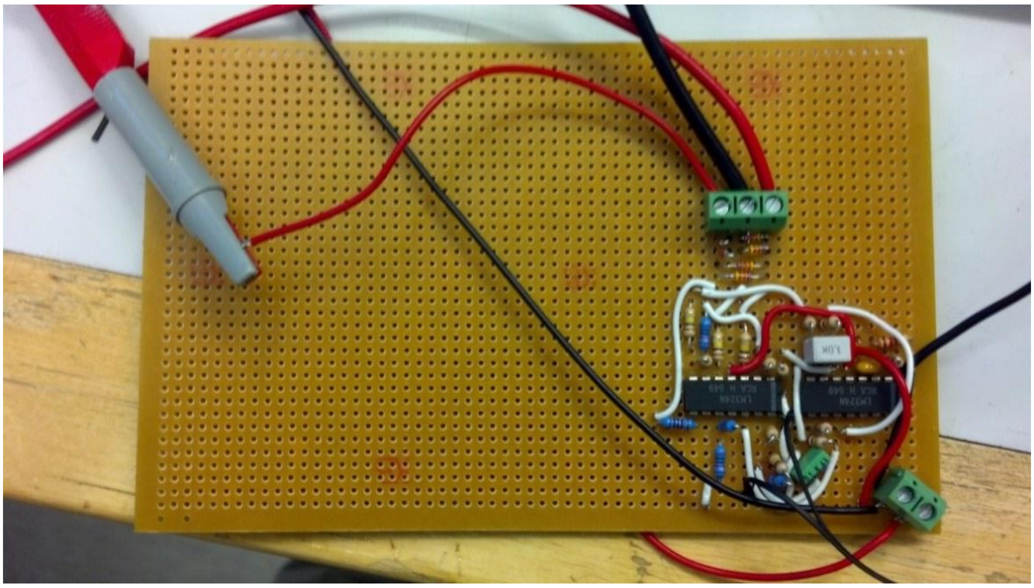


**Figure 28** Manufactured amplifying circuit

The first trial of the system was done using a function generator and an oscilloscope to observe if the system was working, amplifying and filtering adequately. The first success that was observed was the filtering. When frequencies higher than the cut off frequency were applied from the function generator, the signal was greatly attenuated. Then we proceeded with the testing of amplification and the communication with the computer via the headphone jack. During this test we observed that the amplification was working while using the ECG clamp electrodes as we managed to observe the ECG on the computer after some problems with the headset plug that will be discussed at greater extent in the discussion section.

When we knew that the circuit was working properly with tested ECG electrodes in all the stages of the circuit, we proceeded to work on the proposed bracelet electrode. Manufacturing of the bracelet suffered the most deviation from the original plan, since it was not possible to use the Ultraflex conductive fabric tape due to time and economical limitations. Due to not having the proper facilities and the lack of knowledge it was opted to not completely make the bracelet from zero, and it was decided to acquire and use a market wrist support. Having the bracelet, we proceeded to glue the Super-tex P180+AT to their planned locations on the bracelet, as can be seen in Figure 29. The conductive textiles were all placed in a way that they could be accessed by cables while the bracelet was being worn, since the connections from these to the circuit would be via alligator clips.

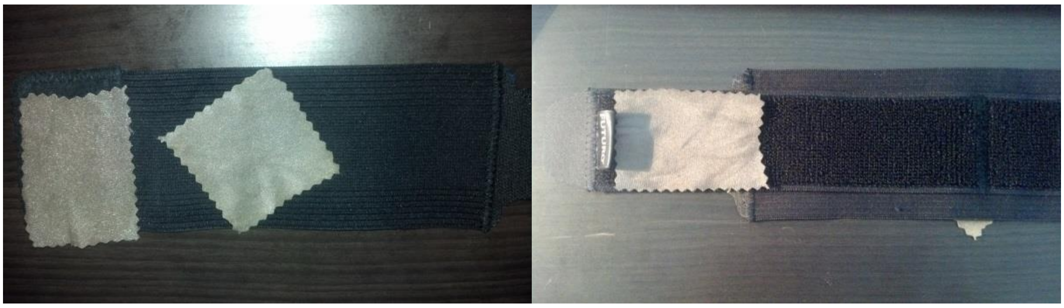


**Figure 29** Manufactured bracelet electrode

# 4 RESULTS

Having the circuitry tested and successfully simulated, we needed to test the rest of the materials that will be used. We did not deem it necessary to proceed to test the textiles of the bracelet, nor the normal hook and loop fastener, especially since it was opted to buy the bracelet and make the necessary adaptations to it. However, we did test the high conductive fabrics, the electrically conductive hook and loop fastener and the conductive fabric tape. The testing of these materials was interesting and various in terms of conductivity, resistivity and tolerance to humidity.

### *Materials*

Firstly we proceeded to check the conductivity of the electrically conductive materials; it was rather straight forward and simple as was the measuring of their resistivity. It was not surprising to see that the characterization was quite similar to those stated in the datasheet for these fabrics (appendixes 2-4). Precisely during these tests we decided on the material to be used for the electrode, which would be the Super-tex P180+AT which had noticeably lower impedance compared to Ultraflex.

All the fabrics were tested against physical factors such as movement, stress, sweat, water and comfort for long term use. They were all found to be resistant to exposure to sweat, repeated washing and were quite comfortable to wear for extended periods of time. This was tested by using the bracelet all day except for sleeping, washing and showering for 10 days. All of them showed impedance lower than 1 ohm, in these specific examples their average was of .417 ohms. We submerged the materials in a saline solution, from which they emerged without suffering any change and then we tried to oxidize the textiles, we also failed in doing so. The washing test was done by loading them along with jeans to a washing machine, repeating the washing cycle on different temperatures, again this test did not cause any change in the properties of the material, neither mechanical nor electrical.

The UltraFlex conductive fabric tape was tested specifically for metal fatigue, since it would be regularly subjected to bending and stress movements. The material was very resistant to the amateur testing performed on it, coupled along with the datasheet’s fact of being able to do a million cycles before succumbing to abrasion; it would be safe to consider that it will be able to withstand the daily use of a bracelet. [21]

Finally it is important to mention that all of these conductive fabrics are compliant with RoHS (Restriction of Hazardous Substances Directive) and REACH (Registration, Evaluations, Authorization and Restriction of Chemicals), however only the P180+AT is classified as a product class I by the standard 100 of Öko Tex. [45] [46] [21]

***Circuit***

Firstly we tested the circuit by checking all the connections utilizing a multimeter. Then we continued to test the circuit we utilized the function generator, the oscilloscope and Matlab. By implementing a sine wave into the system of 10 hertz and amplitude of 1 millivolt and observed the output signal, which can be observed in Figure 30. From this figure we can see that all components are functioning as desired and that the proper testing might give better results than expected.

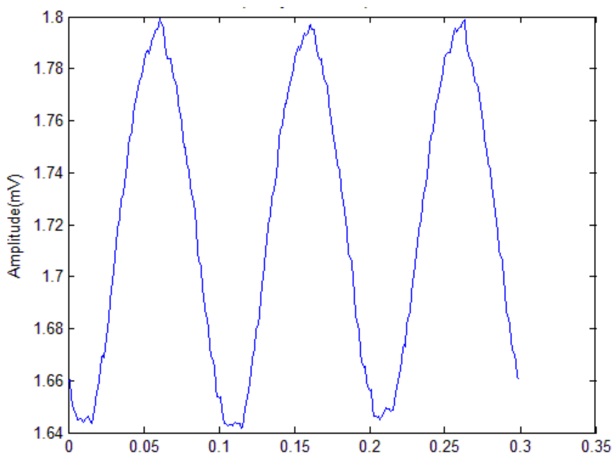



**Figure 30** Sine wave data plot with 10 hertz and 1 millivolt amplitude

***Bracelet***

After having tested the circuit and receiving a positive outcome, it was decided to observe the behavior of the bracelet and the silver electrodes. Firstly we measured that the cables connected to the textile electrodes were working appropriately and were conducting the data. Then we proceeded to connect the bracelet's output into the Biopac

MP36 as can be seen in Figure 31. The analogue input channel of the Biopac MP36, for all the measurements, was set to the preset of ECG (.05-150 Hz).

**Figure 31** Bracelet placement while connected to the Biopac MP36

From this test we managed to obtain a good ECG waveform. However, it had some noise due to the impedance of skin-electrode interface, which got drier towards the end of the test. This noise was easily removed when using the wavelet denoising toolbox in Matlab, the comparison of the original data and the denoised data can be seen in Figure 32. The code used for cleaning the signal can be found in Appendix 1

.

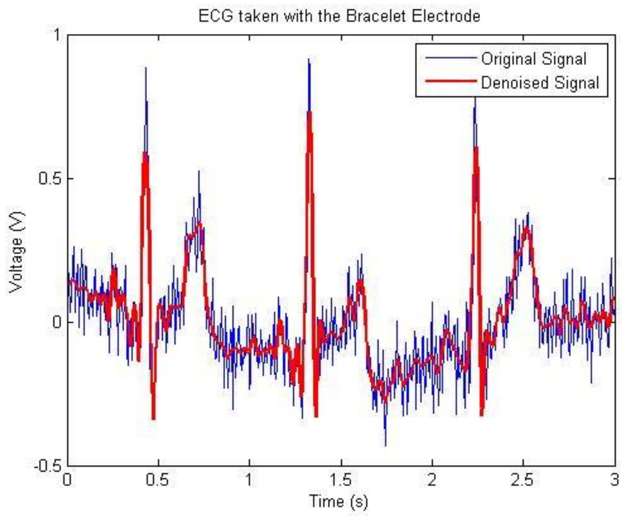



**Figure 32** ECG waveform using bracelet electrode and Biopac MP36.

Having managed to clean the signal, we proceeded to test the bracelet both when the electrodes were humid because of sweat, the results of which can be seen in Figure 33.

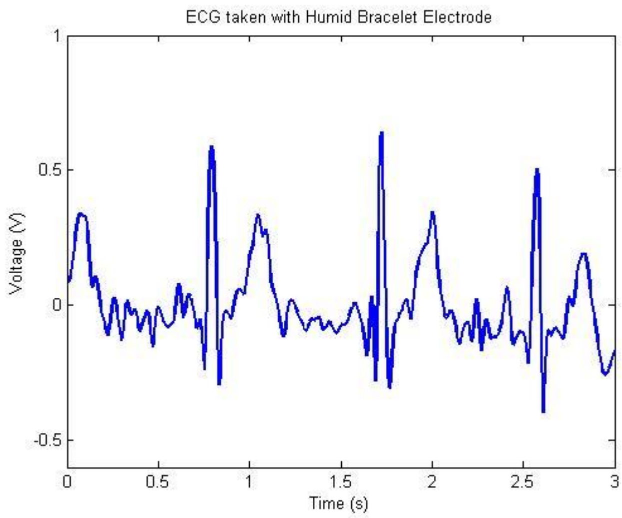
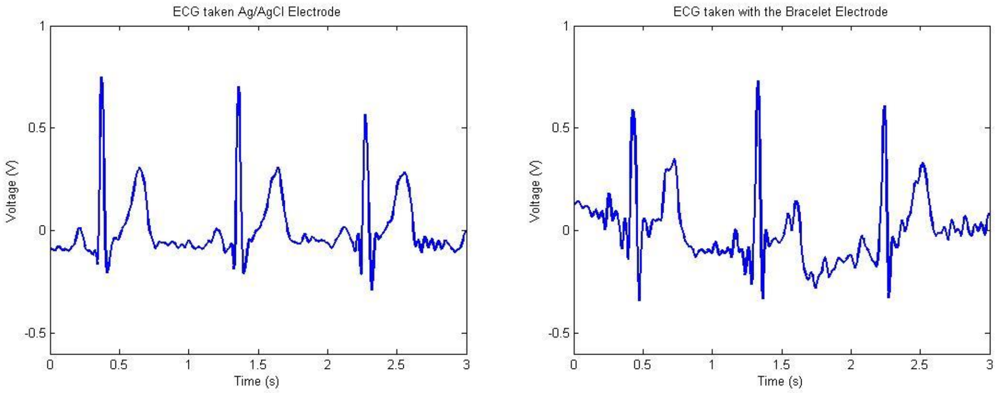



**Figure 33.** ECG waveform taken with humid bracelet electrode and Biopac MP36.

The last test was conducted with the Biopac MP36 was using Ag/AgCl electrodes. Observation from the Ag/AgCl electrodeswas used as a reference signal to compare the ECG waveforms captured from the textile electrodes. The placement for the electrodes can be seen in Figure 34.

**Figure 34** Ag/AgCl electrode placement on left and right hands while connected to the Biopac MP36

The results from both electrodes can be seen side by side in Figure 35. It can be seen that the ECG captured using the bracelet electrode has more noise, though the waveform is acceptable to observe some of the more pronounced cardiac afflictions.




(a) (b)

**Figure 35** ECG waveform taken with both Ag/AgCl (a) and bracelet electrode using the Biopac MP36 (b)

***Bracelet and Circuit***

Having finished the design, the manufacturing, and the testing of the individual components, we proceeded to evaluate the functionality of bracelet for its intended use. For the tests we used the sound card of a Laptop, model Lenovo W520 from Mexico, with the Goldwave 5.58 audio recorder software for capturing the ECG waveform.

We managed to observe the result of the test performed to check the proper functioning of the circuit and communication system. Also we could proceed to record the data received through Goldwave, which has also the option to save the data in Matlab (.mat) format, allowing us to use Matlab for post-processing of the signal. The data obtained in the initial measurement can be observed in Figure 36.

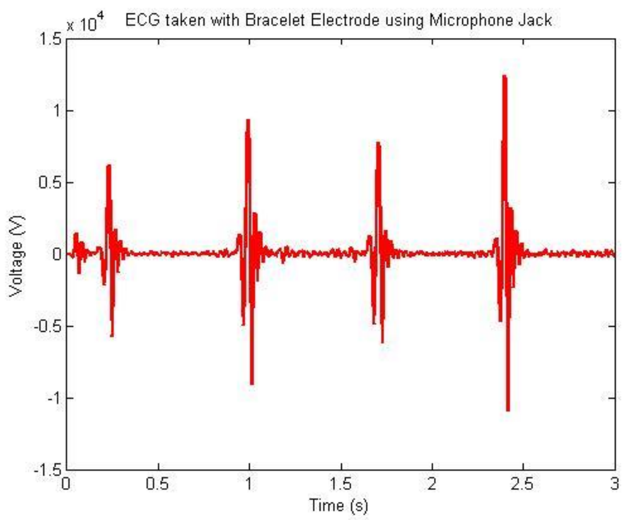



**Figure 36.** ECG waveform taken with bracelet electrode and the proposed communication system before post-processing of the signal on MATLAB.

As it can also be seen from Fig.36, there was certain noise in the signal. However, compared to the simplicity of use and comfort, amount of noise can be considered as acceptable During the 10 day trial of the bracelet use, the bracelet did not prove to be uncomfortable, though this version was not very fashionable. The daily measurements taken during the period were consistent and similar to the one presented here, though there were times that we could observe an increase in motion artefacts.

For using smartphones as a real time audio signal processing device there were problems regarding firmware, specifically in Android that will be discussed more in depth in the discussion and conclusion part of the thesis. However as recording devices they worked quite well allowing us to post process the signals. It is necessary to clarify that there was minimal amounts of motion interference in the signal.

# 5 DISCUSSION

The main focus points of discussion were the post processing which was quite straightforward and simple to do using Matlab and the toolboxes that come with it. However it would be ideal to find better ways to post process the data so as to obtain a better waveform. It is important to remark that the analogue filtering helped to make the post processing easier.

After comparing the signal obtained from the bracelet and the Biopac MP36 versus the ECG waveform that was obtained with the proposed circuit, we concluded that the circuit needs to be revised in order to improve the quality and precision of the components. It would be necessary to augment the amount of analogue signal processing that is being done, by including some of the post-processing methods such as initial noise filtering in the hardware

However, it would be a good idea to increase the area of the electrodes to obtain a better signal acquisition. The idea to add padding to improve contact between the electrode and skin was considered. To test this idea we proceeded to tighten the bracelet, however, we discarded it after not seeing any noticeable improvement in the results, which can be seen in Figure 37.

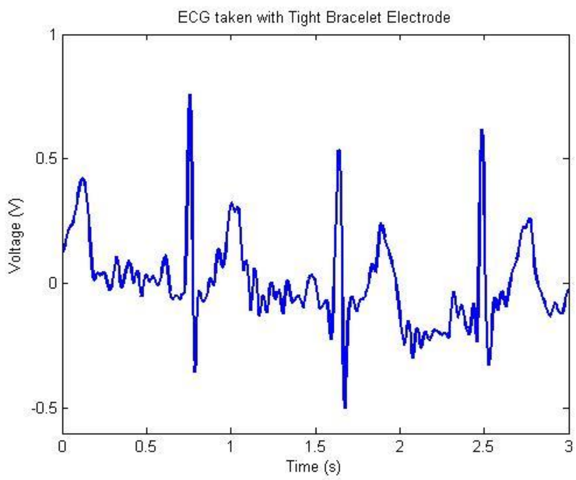



**Figure 37** ECG waveform using the bracelet electrode extra tight.

The main problem we ran into with the communication system which caused a lot of problems and could render the whole system virtually useless was that we ignored the difference between the headset's plug and the microphone's plug schematic. This difference can be seen in Figure 38.

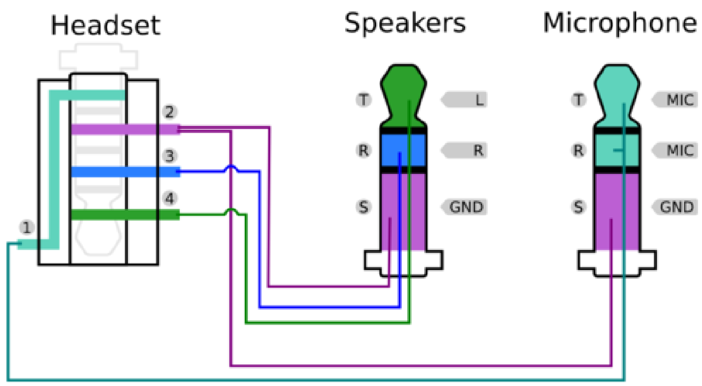



**Figure 38** Headset schematic [47]

The headset schematic was added here due to the fact that until the testing stage did this come to our attention, we were using only a microphone jack. It is very important to use the headset jack since it is essential to make this ECG be functional for mobile phones and digital recorders. Having the opportunity to make this system supported by multiple platforms will allow us to increase the patient's freedom and comfort.

It is important to remark that users of Android phones will only be able only to record due to the fact that their native development kit does not have support for real time low latency audio; with simultaneous and synchronous play and record. [48] However IOS users do have the ability and the applications to work with the signal and view their ECG in real time. Thus it would be recommended to create a specialized application for IOS systems and ECG viewer for Android devices for recorded data.

Ensuring this growth opportunity, it is necessary to make the system battery operated; this system can already work on a 9 volt battery. It would be especially interesting if the battery was rechargeable and the circuit can theoretically be recharged using a connection via the universal serial bus:

We tested using the universal serial bus as a power supply with the prototype and it worked properly. It caused a considerable amount of curiosity on increasing the circuit to handle an analogue to digital converter and using the universal serial bus for the data transfer. However, this would cause a slight increase of size on the circuit and when it would be integrated it might cause the bracelet to be thicker.

Taking into consideration the increasing size of the bracelet, made the choice for the padding relatively difficult due to the fact that it was needed to have a good support and protection without noticeably increasing the volume. Nonetheless it is necessary to comment that at one point it was greatly considered to utilize a conductive fabric that included padding to maximize space. Under careful analysis it was observed that this padding had a more abrasive covering textile than others that we tested.

Another option that would improve the quality of the measurements is to utilize two bracelets and use the electrically conductive hook and loop fastener for intercommunication. This would be mainly to decrease the amount of motion artefact and would be interesting for people whose doctors need to have a less noisy ECG for more precise home care supervision.

There are numerous applications for this system, from following Nymi into the biometric area, to the main reason it was developed, to allow patients to periodically check their ECG. The latter option would be quite interesting to develop because it could be easily implemented on a budget in wheelchairs and hospital beds, using Raspberry Pi's, the credit card sized single board computer, for the more complete version of the system.

# 6 CONCLUSION

In this thesis, different electrode types and fixture alternatives were evaluated in order to propose a more cost-efficient system for continuous ECG measurements with an acceptable level of signal quality. This was achieved with Super-tex P180+AT and a sports bracelet. Measurements, where individual parts were tested for their performance, were conducted for electronics and bracelet materials. Finally whole system was tested with BioPac MP36 system, resulting in sufficient signal quality with 1-lead electrode configuration.

In overall, the bracelet electrode constructed with the selected textile electrode and sports bracelet, functioned as intended.It provided a low cost, easy-to-use alternative for patients who need continuous ECG measurements throughout their treatments. It is useful for both short-term and long-term uses as it provides comparable signal quality to ECGs with more leads. It is important to remark that the ECG obtained using the microphone jack would not be clinical grade, and only some cardiac malfunctions may be diagnosed with it.

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# Appendix 1

```
% ECG

clear;clc;close all;

load('ECG.mat')

%stating Variables

    lev = 2;

    T= linspace(0,3,501);

    T1= linspace(0,3,3000);

%Detrending Signals

    dt_AgAgCl2=detrend(AgAgCl2);

      dt_bracelet100=detrend(bracelet100);

      dt_JLR=detrend(JLR);

     dt_braceletT=detrend(braceletT);

        dt_braceletW=detrend(braceletW);

        dt_braceletbest=detrend(braceletbest);

%Timeframe Selection

    dt_AgAgCl2=dt_AgAgCl2(2000:2500);

    dt_braceletT=dt_braceletT(2000:2500);

      dt_braceletbest=dt_braceletbest(2270:2770);

      %dt_bracelet100=dt_bracelet100(4300:4800);
```

```
    dt_JLR=dt_JLR(1:3000);

    dt_braceletW=dt_braceletW(3400:3900);

  %Removing Mean

mean=mean(dt_AgAgCl2);

dt_AgAgCl2=dt_AgAgCl2-mean;

clear mean;

mean=mean(dt_braceletT);

dt_braceletT=dt_braceletT-mean;

clear mean;

mean=mean(dt_braceletbest);

dt_braceletbest=dt_braceletbest-mean;

clear mean;

mean=mean(dt_JLR);

dt_JLR=dt_JLR-mean;

clear mean;

mean=mean(dt_braceletW);

dt_braceletW=dt_braceletW-mean;

%Final Filtering and Plotting

     AgAgCl2filt = wden(dt_AgAgCl2,'heursure','s','one',lev,'sym8');

   figure;plot(T,dt_AgAgCl2); hold on;
```

plot(T,AgAgCl2filt,'r','linewidth',2), title ('ECG taken with Ag/AgCl Electrode'), ylabel ('Voltage (V)'), xlabel('Time (s)')

legend('Original Signal','Denoised Signal');

figure; plot (T,AgAgCl2filt,'b','linewidth',2), title ('ECG taken Ag/AgCl Electrode'), ylabel ('Voltage (V)'), xlabel('Time (s)'),axis([0 3 -0.6 1])

JLRfilt = wden(dt_JLR,'heursure','s','one',lev,'sym8');

figure; plot (T1,JLRfilt,'r','linewidth',2), title ('ECG taken with Bracelet Electrode using Microphone Jack'), ylabel ('Voltage (V)'), xlabel('Time (s)'),%axis([0 3 -0.6 1])

braceletTfilt = wden(dt_braceletT,'heursure','s','one',lev,'sym8');

figure;plot(T,dt_braceletT); hold on;

plot(T,braceletTfilt,'r','linewidth',2),title ('ECG taken with Tight Bracelet Electrode'), ylabel ('Voltage (V)'), xlabel('Time (s)')

legend('Original Signal','Denoised Signal');

figure; plot (T,braceletTfilt,'b','linewidth',2), title ('ECG taken with Tight Bracelet Electrode'), ylabel ('Voltage (V)'), xlabel('Time (s)'),axis([0 3 -0.6 1])

braceletWfilt = wden(dt_braceletW,'heursure','s','one',lev,'sym8');

figure;plot(T,dt_braceletW); hold on;

plot(T,braceletWfilt,'r','linewidth',2),title ('ECG taken with Humid Bracelet Electrode'), ylabel ('Voltage (V)'), xlabel('Time (s)')

legend('Original Signal','Denoised Signal');

figure; plot (T,braceletWfilt,'b','linewidth',2), title ('ECG taken with Humid Bracelet Electrode'), ylabel ('Voltage (V)'), xlabel('Time (s)'),axis([0 3 -0.6 1])

braceletbestfilt = wden(dt_braceletbest,'heursure','s','one',lev,'sym8');

```
figure;plot(T,dt_braceletbest); hold on;

plot(T,braceletbestfilt,'r','linewidth',2),title ('ECG taken with the Bracelet Electrode'), ylabel ('Voltage (V)'), xlabel('Time (s)')

legend('Original Signal','Denoised Signal');

figure; plot (T,braceletbestfilt,'b','linewidth',2), title ('ECG taken with the Bracelet Electrode'), ylabel ('Voltage (V)'), xlabel('Time (s)'),axis([0 3 -0.6 1])
```

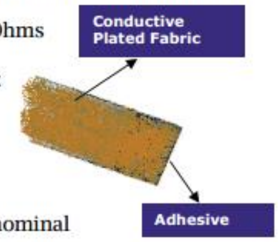
# Appendix 2

## Technical Data Sheet

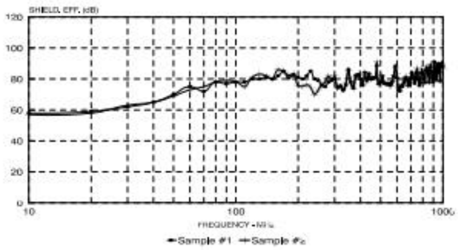
Shieldex® produced by statex

### Conductive Fabric Tape Ultra-Flex™

- **Purpose:** Conductive Fabric Tape for General Use
- **Description:** Tin-Copper Plated PBN-II® Non-Woven Fabric with Conductive Adhesive
- **Surface Resistance:** Maximum 0.09 Ohms - Average 0.05 Ohms
- **Shielding Effectiveness:** Average 80db from 30Mhz to 1Ghz
- **Abrasion Resistance:** 1,000,000 Cycles
- **Temperature Range:** -30°C to 100°C
- **Total Thickness:** 0.127 mm to 0.150 mm (0.005 to 0.006") nominal
- **Weight:** 0.07 to 0.09 Kg/ .84 M² (2.3 to 3.0 oz/square yard)
- **Roll Slit Widths: Cut to size as customer requested**
- **Roll Lengths: 18yds**




UltraFlex Conductive Fabric Tape
Mil-Std 285 Modified

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*A Woman's Owned Small Business*

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**Fax: 315-597-6687**
**Email: whoge@rochester.rr.com**
**Www.shieldextrading.net**

Appendix 3

SHIELDEX® Technik-tex P130+B

# Technical Data Sheet

PN# 1150902130TB

| PRODUCT DATA | Shieldex ® Technik-tex P 130 + B |
|---|---|
| Purpose | High conductive fabric for Smart Textiles |
| Description | Silver plated knitted fabric |
| Raw material | 78% Polyamide + 22% Elastomer |
| Plating | 99% pure Silver |
| Coating (+B) | Polyurethane as additional protective coating |
| | DS (double stretch direction –wrap –weft) |
| Stretch | < 5 Ohms/□ |
| Surface Resistivity | -30˚C to 90˚C |
| Temperature Range | 0.45mm ± 10% |
| Total Thickness | 132cm ± 5cm |
| Roll Width | Average 30m |
| Roll Length | 104g/m2 ± 10% |
| Weight raw | 135g/m2 ± 10% |
| Weight plated and coated | |
| RoHS | Compliant |
| Reach | Compliant |

**PRODUCT APPLICATION**

Base material for technical application

Technical Textiles Inc.

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Alterations reserved
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Appendix 4

Shieldex produced by statex

# Technical Data Sheet

PN# 1101301180

## MedTex™ P-180

- Purpose: Wound Care, Antimicrobial products and Garments
- Description: High Ionic silver release plated nylon elastic knit single direction
- Surface Resistance: Average <5 Ohms/
- Plating: 99.9% pure silver
- Abrasion Resistance: 10,0000 cycles
- Temperature Range: -30c to +90c
- Total Thickness: 0.55mm
- Weight: 224 g /m²
- Stretch: Double stretch direction (wrap and weft)
- Roll Lengths: 50 LY average
- Roll Width: 135cm
- 78% Nylon + 22% elastormer

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# Appendix 5

Texas Instruments

LM124-N, LM224-N, LM2902-N, LM324-N

www.ti.com

SNOSC16B – MAY 2004 – REVISED SEPTEMBER 2004

## LM124-N/LM224-N/LM324-N/LM2902-N Low Power Quad Operational Amplifiers

Check for Samples: LM124-N, LM224-N, LM2902-N, LM324-N

### FEATURES

- **Internally Frequency Compensated for Unity Gain**
- **Large DC Voltage Gain 100 dB**
- **Wide Bandwidth (Unity Gain) 1 MHz (Temperature Compensated)**
- **Wide Power Supply Range:**
  - **Single Supply 3V to 32V**
  - **or Dual Supplies ±1.5V to ±16V**
- **Very Low Supply Current Drain (700 μA)—Essentially Independent of Supply Voltage**
- **Low Input Biasing Current 45 nA (Temperature Compensated)**
- **Low Input Offset Voltage 2 mV**
  - **and Offset Current: 5 nA**
- **Input Common-Mode Voltage Range Includes Ground**
- **Differential Input Voltage Range Equal to the Power Supply Voltage**
- **Large Output Voltage Swing 0V to $V^+ - 1.5V$**

### UNIQUE CHARACTERISTICS

- **In the Linear Mode the Input Common-Mode Voltage Range Includes Ground and the Output Voltage can also Swing to Ground, Even Though Operated from Only a Single Power Supply Voltage**
- **The Unity Gain Cross Frequency is Temperature Compensated**
- **The Input Bias Current is also Temperature Compensated**

### ADVANTAGES

- **Eliminates Need for Dual Supplies**
- **Four Internally Compensated Op Amps in a Single Package**
- **Allows Directly Sensing Near GND and $V_{OUT}$ also Goes to GND**
- **Compatible with All Forms of Logic**
- **Power Drain Suitable for Battery Operation**

### DESCRIPTION

The LM124-N series consists of four independent, high gain, internally frequency compensated operational amplifiers which were designed specifically to operate from a single power supply over a wide range of voltages. Operation from split power supplies is also possible and the low power supply current drain is independent of the magnitude of the power supply voltage.

Application areas include transducer amplifiers, DC gain blocks and all the conventional op amp circuits which now can be more easily implemented in single power supply systems. For example, the LM124-N series can be directly operated off of the standard +5V power supply voltage which is used in digital systems and will easily provide the required interface electronics without requiring the additional ±15V power supplies.