Reader for the assignment:

**Electrical measurements on the human body**

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**In Progress**

1. Add calculations of gains and filters of circuit.  
2. Add description of biofeedback measurement configurations: EOG, EMG, GSR
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1 Introduction

This is the reader for the assignment ‘Electrical measurements on the human body’ and since you are reading this document you have chosen to attain to this assignment. Probably you had some reasons to do this, but let’s emphasize once more the importance of medical healthcare and the place it has in our society. In our welfare state the elderly are becoming older each year. This is possible thanks to the increased quality of medical care during the last few decades. Although the main subject of this assignment, recording of an ECG, is already a very old principle it is still of utmost importance for a cardiologist to for example diagnose many heart diseases. Besides its application in healthcare many other applications of ECG exist. Most of these other applications can be found as sport applications. For example fitness equipement which also monitors your heart rate, so the user has some feedback of his efforts and can be warned in case of extremely deviant heart rates. Of course many other applications can be found, think of example of ECG as a measure of relaxation, making it very applicable as a lie detector. This assignment will enable you to incorporate human parameters in a product; providing you even more design possibilities. The reason to choose ECG for this assignment, which might be considered a little old-fashioned, is because the signal is relatively easy to measure, without the need for sophisticated medical equipment. To record ECG knowledge of the human body, electrical circuits and even digital signal processing is required, which makes it a very interesting and educational subject.

In principle this document should provide a complete image on the bioelectricity in the human body, how to measure it and which information it holds. Detailed information is omitted, but curious students are always encouraged to follow the web links or to search for the keywords provided after each paragraph.

All subjects will be discussed during three meetings. After these meetings there will be two more weeks to work on your own ECG application. The structure is as follows: first the basic principles will be discussed. This involves nerves cells and their operation. Secondly the human heart will be discussed. Finally measurement principles and safety issues will be discussed. If time permits the brain will be discussed in the third week as well.

The last two weeks will be completely different. After the third week you will be asked to present your design ideas to the group after mailing it to the lector. Presentations can be held as a couple or individually. The rest of the students attaining to the assignment will be asked to provide feedback on your ideas. Think for example of questions like: will it work? Is your idea innovative? Which values do you extract from the ECG? Etc. If the assignment lector is satisfied with of the level of your proposal you can continue to build a prototype, your task for the last two weeks. At the end you have to write and hand in a short report and demonstrate the prototype.
2 Nerve cells and their operation

Nerve cells are the building blocks of the nervous system, which consists of the spinal cord, brain and many peripheral nerves. Nerve cells, often called neurons, are the cells in living beings that make information transfer from one side of our body to the other possible. As the title of this assignment already suggests the neurons use electric properties to process and transfer information. How this works will be explained step by step in this first chapter.

2.1 Anatomy of nerve cells

Neurons share a lot of characteristics with other animal cells; besides these characteristics they posses some electric properties, which enables information transfer. To keep this reader a little condensed only the electrical properties and some other, as far as they are required, are discussed. The neuron consists of a large body, the cell body or soma. This part of the cell includes the nucleus and all other cell systems. From the soma small branches, called dendrites and axons, rise. Dendrites are the branches that carry information to the soma, while axons always transfer information to other parts of the nervous system or muscles. In other words dendrites are the input of the neuron while axons are the output. A very simplified picture can be found in figure 1.

![Figure 1, simplified nerve cell](image)

In most cases every neuron has only one axon; however this axon can undergo many bifurcations and reach for example a muscle over a distance of more than one meter. Think for example of a neuron in the spinal cord near the brain innervating a muscle near one of the feet. To make a fast communication possible for those long axons, they are covered by myelin sheaths, which are created by Schwann cells. The advantage of these sheaths will be explained later.

2.2 Resting potential

The most important electric characteristic of a nerve cell is its resting potential over its cell membrane. This means that the cell maintains a voltage over its entire membrane as long as there are no stimuli. Under normal circumstances this potential is approximately -75 mV, which means the inside of the cell is more negative than the outside of the cell. The potential difference is caused by concentration differences of ions dissolved in the plasmas inside and outside the cell. In most human plasmas sodium (Na\(^+\)), potassium (K\(^+\)) and chloride (Cl\(^-\)) ions can be found. The cell membrane is semi-permeable, which means small molecules, like water, but also ions of potassium, sodium and chloride, can pass through it with some resistance. For the most important ions, those of potassium and sodium, there are voltage dependent channels in the membrane.

An active mechanism in the cell maintains concentration differences by moving K-ions out of the cell and Na-ions into the cell. Therefore the K-ion concentration is larger outside the cell, while the Na-ion concentration is larger inside the cell. Negatively charged chloride ions are mostly outside the cell. Inside the cell many large negatively charged molecules, which are unable to pass the membrane, compensate for this. Although there is a potential difference, concentrations of negative as well as positive charged ions are almost equal, since an unnoticeable concentration difference already causes the potential difference. During rest ion concentrations as well as the potential difference are kept constant by the cell.


2.3 Action potential, Hodgkin cycle

The nerve cell uses the concentration differences to communicate with other parts of the nervous system. By certain stimulations the resting potential can be disturbed to either higher or lower values, called respectively depolarization and hyperpolarization. Depolarization can cause a so called action potential. Once a certain threshold is reached by the stimulus an active mechanism decreases the potential difference by opening channels in the membrane for the sodium ions. Sodium ions will flow into the cell, causing the potential to rise even further. This mechanism is known as the Hodgkin cycle. Secondly channels for sodium close, while more potassium channels open, causing again a decrease of potential. The potential difference across the cell membrane as a function of time is depicted in figure 2.
After an action potential the concentrations and the resting potential on both sides of the membrane are restored by the cell. During this period the cell is unable to receive any input at the beginning and later very insensitive to new input, called respectively the absolute and relative refractory period.

Once an action potential is started a certain point on the neuron membrane surrounding this place will be depolarized as well due to passive diffusion of the ions. The passive depolarization causes the membrane voltage to reach the threshold value starting again the active cycle. This way the action potential is carried along the whole cell membrane reaching a speed in the order of a few meters per second. For a human being this would mean that it takes approximately one second before you notice someone is standing on your toe. Luckily this is not the case.

Further reading: Wikipedia – Action potential

2.4 Schwann cells, Myelin

As mentioned before most long axons are covered by myelin sheets. Between every sheet there is a small piece of uncovered axon, called the node of Ranvier. The sheets surround the entire axon and are made of a fat like substance which has no free ions.

An action potential is unable to pass under the myelin sheet; instead the passive ion flow reaches the next node of Ranvier causing an action potential at that node. In other words the action potential jumps from node to node, increasing the propagation speed by approximately 20 times.

2.5  **Neurotransmitter, Post Synaptic Potential**

Up till now we only discussed the propagation of information in a single neuron; however the nervous system consists of millions of these neurons. Therefore there has to be a way to transfer the action potential to another nerve cell. The way this is done is by neurotransmitters, a substance which is secreted by a neuron and influences the permeability for different ions of another neuron.

A synapse is placed close to the cell membrane of another neuron, as can be seen in figure 3, or the cell membrane of a muscle cell. In this figure the synapse is located close to a dendrite, but this can also be the soma or even the axon of the second neuron. The neurotransmitter is secreted at the extremities of an axon, called synapses, into the synaptic cleft, a very small cleft between the synapse and the other neuron.

![Figure 3, connection of two neurons](image)

The effect of the neurotransmitters is divided in two groups. The first are neurotransmitters which increase the potential difference across the membrane by increasing the permeability for all ions (see figure 4), bringing the potential closer to the action potential threshold. These neurotransmitters are classified as excitatory, causing an EPSP (excitatory post synaptic potential) at the next neuron.

The second are neurotransmitters which do exactly the opposite. Instead of increasing the potential they decrease it to an even lower value by increasing the permeability of the membrane for potassium and chloride ions only. These neurotransmitters are called inhibitory, causing an IPSP (inhibitory post synaptic potential) at the next neuron. A neuron always secretes the same neurotransmitter and always causes EPSP’s or IPSP’s, although some neurotransmitters work as excitatory as well as inhibitory transmitter.
Although a single EPSP causes the membrane potential to rise, they seldom cause an action potential. In the synaptic cleft enzymes break down the neurotransmitter and stop the membrane potential increase. At the same moment the neuron tries to restore the resting potential. However an action potential can be achieved by more than one EPSP. Temporal and spatial summation can cause the potential to reach the threshold value.

A single neuron receives input from many other neurons whose synapses are sometimes located close to each other. The cell membrane passively conducts an EPSP, by ion diffusion along the membrane, increasing the membrane potential for nearby synapses as well. This decreases the difference between the resting potential and the threshold value, making it easier to trigger an action potential for a nearby EPSP’s. Of course the same holds for an IPSP.

A single PSP lasts much longer than a single action potential. This will lead to a summation of the PSP’s in time. For example the membrane is still less polarized from the last EPSP while a new EPSP arrives. So if enough EPSP’s reach a single spot on the membrane an action potential will be the result.

Drugs and alcohol mainly influence the communication by neurotransmitters. Some of them block excitatory receptors while others block inhibitory receptor. Caffeine for example blocks the inhibitory neurotransmitter Adenosine, which is believed to promote sleep and suppress arousal. The caffeine molecules resemble the adenosine molecules enabling them to block its receptors, which cannot be triggered anymore to change the membrane permeability. The user feels less tired and is more sensitive to arousal.

Further reading: Wikipedia – Chemical synapse
3 Equivalent electrical networks

To gain a better insight in the electric information transfer of neurons one could model them. Instead of ions in a solution this is done by electrodes in electronic components like resistors and capacitors. Two different models are distinguished: firstly a simple one in which there is no action potential and secondly one which (partly) incorporates the action potential.

3.1 Model for passive conduction

The simplest form of modeling a neuron is to only model the passive conduction at the membrane. Since the membrane separates two solutions it can be described as a capacitor. The ions in both solutions cannot move completely free but experience some resistance; therefore we place a resistor between the nodes. The resistance for ions inside the cell is much larger than the resistance for ions outside, because of the small diameter of the axons and dendrites, which makes it possible to neglect the resistance outside the cell. This model already provides a lot of possibilities however the resting potential is still not incorporated. Adding a simple voltage source in series connection with a resistor models this neuron property. The model for three nodes can be found in figure 5.

![Neuron Model](image)

Figure 5, neuron model in case no action potential arises.

Suppose we apply a current to the first node of the model, then the voltage at that node will change accordingly to the direction and strength of the current. Simple network theory like applying a KVL reveals that the equation describing the voltage across the first capacitor is a differential equation. Thus the current will charge the capacitor with an exponential curve which time constant depends on the values of $R_{ax}$, $R_m$ and $C_m$.

The fact that the myelin sheets created by the Schwann cell increase the transmission speed of the action potentials can be explained by this model. It can be incorporated in the model as a decrease of the capacitance $C_m$, since myelin is a fat like substance. A capacitance depends on the covered surface and intermediate distance. The myelin sheets increase the distance of the plasma inside and outside the cell, effectively decreasing the capacitance.

The applied current also causes a voltage change at the second node. If we consider the nodes to be infinitely small the change at a distance $x$ behaves also exponential to the applied current at the input, however the effect decreases very rapidly with increasing distance $x$. Therefore the active conduction is needed to transfer the information over some distance.
3.2 Model including the Hodgkin cycle

The active conduction is done by the mechanism described earlier: the Hodgkin cycle. Once a certain threshold is reached, the membrane becomes permeable for sodium ions, causing a large influx of these ions which raises the membrane potential even further. This can easily be modeled by a comparator sensing the membrane potential followed by a one-shot (a circuit that generates a single pulse), which models the temporary permeability for sodium. This can be seen as switching to another potential \( E_{\text{na}} \), the resting potential for a membrane only permeable for sodium ions. The model for one node is depicted in figure 6.

![Figure 6, neuron model in case an action potential arises.](image)

3.3 Assignment on building a neuron model

The last two paragraphs discussed how a nerve cell could be modeled. These models, and especially the first model, are very suitable to be build by using electrical components and a bread board. Students willing to gain more insight in propagation of membrane potentials can build such a model, containing for example three lumps. However this is an optional task and not part of the regular course.

First try to estimate which values to use for the components. Maybe you would like to model the membrane as faithfully as possible, however this yields a very fast circuit. Therefore it would be wiser to model is as a rather ‘slow’ circuit which will enable you to measure some potentials over time.

Secondly there is a challenge to visualize the potentials. Just connecting for example a LED at each lump will certainly not work, since LED’s require much more current than available in the circuit. Think of something which measures potential without the need for (much) current. Modeling the active circuit will be the biggest challenge, since is involves the need for a comparator and one-shot circuit. Students willing to build this circuit can contact the assignor to find the appropriate components.
4 Potentials at the skin

Up to this point only the anatomy and modeling of nerve cell inside the human body has been discussed. However we do not want to use needless or other equipment that needs to be brought into the body to measure for example an action potential. Instead electrodes attached to the skin are used to measure electric activity. Metal electrodes attached to the skin do not measure the membrane potential difference and do not use ions, but electrons to generate a current. The next sections briefly explain what is measured and what the consequences are of this method.

4.1 Volume conduction

An action potential generates ion currents. Especially sodium, potassium and chloride ions are involved in an action potential. Consider a nerve cell in which an action potential passes while the potential changes are observed from a short distance. A very simplified action potential happens in two phases: a depolarization and a repolarization phase. During depolarization sodium ions flow into the cell while during repolarization potassium ions flow out of the cell. These two flows can be considered respectively as a current sink and source, which is depicted in figure 7.

The extracellular solution can be considered as a resistive medium, when a current is applied to this medium there will be a potential difference between the place of the source and sink of the current. Ion current will spread through the whole medium, which is considered to be homogeneous, causing a changing potential over time when an action potential passes by.
Suppose we measure the potential at a distance $a$ (see figure 7) from the cell membrane compared to a potential far away, which we assume to be zero. Since the action potential moves from left to right, it would be the same if we measure the potential from right to left over the line at distance $a$ from the cell membrane. With increasing time firstly the potential decreases to a minimum just above the sink then it passes zero to reach a maximum just above the source.

The waveform discussed is related to the depolarization front inside the neurons axon, as you can imagine the repolarization front will cause a similar but opposite waveform at the same time at a short distance. An electrode attached to the skin close to the neuron will therefore measure a summation of both phenomena.

Although this is a very simplified model the signals measured in clinical set-ups closely resemble the wave shapes derived from the model. During more practical set-ups, like the one used later during the assignment, not a single action potential of one neuron but a summation of simultaneous action potentials of multiple neurons will be measured. The discussed theory is to illustrate why it is possible to measure potentials at the human skin.

*Further reading:* Wikipedia – extracellular field potential

### 4.2 Electrodes

As already discussed before the human body generates potentials and currents by means of ions in solution. The mental electrodes used to measure bio-electricity use electrons to generate potentials and currents. Therefore there has to be some kind of mechanism to convert the ion currents to electron currents. This mechanism can be found at the skin-electrode transition.

When a metal is brought into water, some atoms will break out of the grid and become ions. Due to the transition from normal atom to ion there will be a voltage difference between solution and metal. The voltage difference causes the charged ions to fall back into the metal grid. After some time there will be equilibrium between the atoms leaving the grid and the ions which are forced back into the grid. The voltage difference caused by the ions is called electrode bias and as can be read in the next section this bias has some major consequences.

In most cases it is already sufficient to put a metal to the human skin to be able to measure the electric signals from inside the body. A transition can be improved by adding a solution with many dissolved ions between the electrode and the skin. For the ECG we can for example use a sponge with a solution of normal kitchen salt, but many more professional ‘electrode pastes’ are available.

*Further reading:* Wikipedia - Standard electrode potential

### 4.3 Signal of the electrodes

When applying these electrodes one is able to record signals for electrical activity. In most cases this activity is generated by nearby nerve cells. Note that an electrode will never record the signal of a single nerve, but in general the activity of many cells nearby, which includes muscle cells, will be recorded. The electrical activity of the heart is much stronger than all the other signals, which makes it possible to record an ECG between almost all extremities. The location of the electrode is not very important to record an ECG. An electrode can be applied to the shoulder as well as to the wrist. A typical ECG is in the order of a few millivolt, while for example EEG activity is in the order of 100 microvolt.
4.4 Artefacts

An artefact is an unwanted part of the signal, which sometimes makes it impossible to read the preferred signal. Artefacts can be caused by many different sources. Most prominent are the artefacts caused by movements, other bio-electric sources and electric mains.

One of the most important implications of the electrode bias is the existence of movement artefacts. The dissolved ions are not able to move very easily in the solution. This means that the dissolved ions close to the electrode will stay behind when the electrode moves over the skin, causing a relatively large voltage at the electrode. Every metal has a different equilibrium and therefore some metals are relatively insensitive to movements, while others are very sensitive. Stainless steel for example is very insensitive, while aluminum is very sensitive. A very obvious solution is of course to choose metal which is very insensitive to movements and to fix the electrode as stiff as possible.

Besides movement artefacts one can also experience artefacts by other bio-electric sources. A clear example is the electric signals generated by activated muscles. An ECG recording for example can be seriously disturbed by electricity from muscles in the breast. Eyes can cause serious problems during EEG recordings. The front of the eye is positively charged compared to the retina and is discharged during a blink, therefore every blink will cause a strong fluctuation in the very small EEG signals. Especially the signals measured at the front of the head experience this problem.

A major consequence of our electric mains it that there is almost no place anymore which is free of mains interference. When measuring very small bio-electricity signals one will always pick-up a 50 Hz part in the signal. The frontend uses a very old solution to suppress mains interference. Assuming the mains interference is present on all wires it suppresses the common part while only amplifying the difference. This is known as a differential (instrumentation) amplifier with a high common mode rejection ration. Chapter 6 of the reader discusses this and other functions of the frontend.

Further reading: ECGpedia – basics (Artefacts)
5 The human heart

It is well known that the heart is a pump for the blood, but how the pumping action is generated is unknown for most people. First the hydrodynamic properties of the heart will be discussed and later the electric properties, underlying the heart rhythm.

5.1 Anatomy and functioning

The heart consists of two sides each providing a pump for a separated circulation in the human body. The right side of the heart receives blood from many organs except the lungs and pumps the blood through the pulmonary artery to the lungs. At the left side of the heart this is just the opposite. Blood from the lungs is pumped through the aorta to the rest of the body. Note that left and right correspond to the patient and not to the way the heart is mostly depicted. Both places where the blood enters the heart are called atria. From the atria the blood is pushed into the ventricles, which are surrounded by strong muscle tissue. During a contraction of the ventricles’ muscles the blood is forced to flow into the arteries. Between atria and ventricles and arteries valves, which stop the blood from flowing back, can be found. A very simplified drawing of the heart is given in figure 8.

![Figure 8, simplified drawing of the human heart](image)

Further reading: Wikipedia – Heart, Wikipedia – Circulatory system, Animation of the heart, How the heart works, an animated tutorial

5.2 The heart muscle

To be able to understand the functioning of the heart muscles one first has to known some properties of muscle cells. Especially electrical properties of muscle cells closely resemble those of nerve cells. Under normal circumstances the membrane of a muscle cell is negatively charged just as the membrane of a nerve cell. When a muscle is innervated the membrane discharges to a
positive value while contraction takes place. Instead of potassium and sodium ions, calcium ions play an important role in action potentials of muscles and the shape of an action potential differs somewhat from those of nerve cells (see figure 9).

There exists an important difference in shape of the inner (Endocardiac, near the chambers) and outer (Epicardiac, near the outside of the heart) cardiac muscle cells. The inner muscle cells are innervated first and remain depolarized longer compared to those close to the outer surface of the heart. As described later this mainly determines the shape of an ECG. Just as in nerve cells most muscle cells can be stimulated by neurotransmitters at a synaptic cleft. Here a post synaptic potential always causes an action potential on the membrane of the muscle cell. The heart muscle, often called cardiac muscle, however is not innervated by nerves, but generates and conducts action potentials by itself.

Further reading: Wikipedia – Cardiac muscle

5.3 Intercalated discs

Conduction from one cardiac muscle cell to the next is done by intercalated discs. Intercalated discs contain gap junctions, which electrically bind two muscle cells. Gap junctions can be seen as small pipes running from one cell to the other. Only small molecules can pass through them, which is enough to transfer the action potential. Due to the gap junctions in the intercalated discs an initiated action potential runs over almost the entire heart. However the top and bottom part of the heart are electrically separated. In other words the atria are electrically coupled and the ventricles are coupled, but there is no direct connection from the atria to the ventricle cardiac muscle cells.
5.4 Myocyte contraction, SA node

Cardiac muscle cells have another special property compared to normal muscle and nerve cells. The resting potential spontaneously decreases till the threshold for an action potential is reached. This is done by gates which act opposite to those of the action potential. They are triggered to open for sodium and calcium by hyperpolarization instead of depolarization. After the action potential the membrane is polarized till the normal resting potential on which the decrement starts again. Once a single cell reaches the threshold it will also trigger its neighboring cell by means of the gap junctions, thus the cell which depolarizes fastest initiates the complete action. Cardiac muscle cells are also known to have a refractory period which disables them to be triggered just after a contraction. Although all cardiac muscle cells share the spontaneous depolarization property the fastest depolarizing cells normally lay in a small part of the atria, called the SA (sinoatrial) node. As can be seen in figure 11 the SA node is located on top of the right atrium.
5.5 AV node and His bundle

As already mentioned before, the muscles of atria and ventricles are electrically separated. This is done to ensure efficient pumping. When an action potential would run from the SA node immediately down to the muscles surrounding the ventricles the blood would be pushed down instead of up to where the arteries (aorta and pulmonary artery) originate from. It is clear that the action potential should come from below to push the blood upwards. Besides direction timing is important, it takes a while for the blood to flow from atria to ventricles, therefore the upcoming action potential needs to be delayed before it hits the ventricle muscle cells. The just described properties are realized by the bundle of His and AV (atiroventricular) node respectively. After the action potential has depolarized the atria it reaches the AV node which is located on the wall between the left atrium and ventricle (see figure 11). Here the action potential is delayed before it is conducted down by the bundle of His, which runs down between both ventricles and bifurcates to a left and right bundle. At the end of the bundles there are many bifurcations which all lead to muscle cells. Once the muscle cells at the bottom are stimulated the action potential will be conducted upwards over the muscle cells itself.


5.6 Electrical measurements on the heart

It is clear that the heart uses many electrical pathways to generate the pumping action. As we will see in the next paragraph all complexes in the ECG can be related to some mechanical action of the heart, which makes the ECG very useful to study the mechanical actions of the heart.
The heart is a very complex organ to model electrically, especially when one would try to find an exact description for the potentials at the skin using the theory of paragraph 4.1. However there is a relative simple way, proposed by Einthoven in 1924, which describes all electric phenomena. This way the heart is modeled as a single current dipole in the middle of a ball (our chest). Arms and legs form an equilateral triangle on a frontal intersection (see figure 12). One can understand that this model discards a lot of preconditions; however it gives a relative good explanation for the signals we are measuring.

The direction of the cardiac vector, which can be explained by the turning of the heart, in this plane is approximately pointing left outside of the left leg. Vector theory tells us that the effect of this cardiac vector on a vector from right to left arm can be calculated by a projection (figure 12). A measurement between right and left arm is known as standard derivation I. For this measurement a differential amplifier is needed with the positive input connected to the left arm and the negative to the right arm. Although the right leg is not exactly neutral, because it originates not that far from the left leg, it serves as common electrode for the amplifier. More about the amplification can be found in section 7.1.

Other commonly used derivations, standard derivations II and III are between right arm and left leg and left arm and left leg respectively. These derivations produce different outputs, since the projection of the cardiac vector is different for these derivations. A summation of the three derivations, with a negative sign for derivation II, should be zero: $I - II + III = 0$. The minus sign is caused by the esthetical choice of Einthoven to have the largest peak (QRS complex) in the signal pointing upwards.

Over the years many different derivations have been proposed. Nowadays we are mostly using a 12 lead ECG which includes the three first derivations proposed by Einthoven. Since the first three derivations are the easiest to explain and to measure under all possible circumstances we restrict ourselves to them.

Further reading: Wikipedia – Electrocardiogram, ECGpedia – Basics (The ECG electrodes)
5.7 Origin of P, Q, R, S and T in the ECG

An ECG has a typical shape which contains a few remarkable peaks. The peaks (positive as well as negative) are marked by the letters P, Q, R, S and T (figure 13), each corresponding to a certain mechanic action of the heart. The letters peaks of P, Q and R are often referred to as the PQR-complex.

A typical ECG starts with a P top which originates from the depolarization of the atria. This top is not very large and can therefore easily be missed, especially when there is a lot of interference in the signal. The repolarization of the atria is not visible in the ECG because it is a very small signal which completely coincides with the QRS-complex.

Next is the QRS-complex which is the largest part of the ECG under normal conditions. The shape of the QRS-complex can differ significantly, depending on electrode location, derivation and person. Therefore you do not have to be afraid if your own ECG looks a little different. The main reason for the steep edges and the presence of the QRS-complex in general is the fact that the inner muscle cells of the heart are innervated a little earlier than those on the outside of the heart (figure 9). Between the moments the first and last muscle cell to depolarize the large Q peak emerges in the signal.

The origin of the T top is caused by the repolarization of the ventricles, which is strong enough to emerge in the ECG. Again the shape is largely determined by the not simultaneously acting muscle cells of the ventricles (figure 9). During the depolarization phase the repolarization is carried out the other way around: from outside to inside. This also explains why the peaks have the same polarity (something which has puzzled many scientists for a long time).

Further reading: Wikipedia – Electrocardiogram, Hurst J.W. Naming of the waves in the ECG, with a brief account of their genesis (Full text HTML, Full text PDF)
6  The human brain (optional)

Besides measurements on the human heart, there is two other commonly recorded electrical signals. One is the measurements of electric activity on the membrane of muscle cells in general; this is known as electromyography (EMG). The other is the electric activity which is associated with brain activity. The human brain consists of millions of nerve cells, which enables human to have higher cognitive functions like consciousness, emotions and feelings. The next chapter will discuss the general anatomy of the human brain and why and how it is possible to record electric activity as an electroencephalogram (EEG), which is recorded at the scalp.


6.1 Applications of the EEG

Recording of an EEG has both clinical as well as research applications. Think for example of monitoring the EEG during anesthesia and intensive care. These yields a possibility to monitor the level of sedation, since drugs directly influence the EEG, to assess the right amount of anesthesia and to detect potential harm to cognitive and neurological functions. EEG can also be used as a tool to diagnose some mental abnormalities. Often ERP (event related potential) are used to determine if the brain responds correctly to certain stimuli. Thanks to EEG many studies on the human brain were possible, which provided the numerous things known about the brain. The functions of certain areas and frequency bands, discussed in the next sections, are just an example of the things known today. Although there are most sophisticated methods like fMRI (functional magnetic resonance imaging) available today, EEG still has many advantages. EEG can be recorded with a small amplifier and computer and it has a very high temporal resolution compared to fMRI. On the other hand fMRI enables the researcher to investigate the functioning of deeper brain structures as well, while EEG does not.


6.2 Neurons in the cortex

First some very important properties of the neocortex, the outer layer of the brain, will have to be discussed. There are many different cells in the neocortex, some of them relaying signals to the deeper structures in the brain, others to cells which are also in the vicinity or distant in the neocortex. A certain function is never implemented by a single cell; there are always other cells around who have to same function, inputs and outputs. This redundancy makes the brain robust against damage and loss of cells. The pyramidal cells, which enable us to record an EEG, are quite large and are located perpendicular to the scull. The axons of the pyramidal cells run down to deeper structures in the brain or to another group of (pyramidal) cells in the neocortex, but at a different location. The largest dendrite of the pyramidal cell runs to the surface of the neocortex. Inputs to the cells can be divided in six layers and the axons stimulating the dendrite in that layer all originate from the same location. Figure 14 shows an overview of the pyramidal cells in the neocortex.

6.3 PSP as electrical source

When an electrode is placed over one of those pyramidal cells it is possible to measure electric activity. However this electric activity is not generated by action potentials in the cells. A single action potential lasts very short and is a very local phenomenon. All outputs of a single system are located in the same layer of the neocortex and a population is aroused simultaneously, therefore one will measure the summation of all local PSP activities instead. In spite of the fact that a PSP is much smaller than an action potential it is the PSP that is the underlying source of the EEG.

An EPSP in layer 2 for example will cause an influx of positive ions, to compensate for this local change there will be an outflow of ions elsewhere on the cell membrane. Since layer two is almost the top layer most of the outflow will be in the lower layers, causing a positive charge, while the local inflow itself causes a local negative charge. These local charge accumulations cause a voltage change on the electrode just above the cells. Figure 14 depicts this event for an arousal in both layer 2 and 4. For an IPSP the signs are just opposite.

6.4 Known functions of some areas

Since the human brain is under study for a long time already many things are known. Specific functions of most areas are known for example. The first researcher to record an EEG (the German psychiatrist Hans Berger in 1924) already discovered a significant change of patterns at the utmost back of the head when the subject closed his eyes. These patterns change from irregular to very rhythmic when the eyes are closed. Later it was discovered that this area is the primary visual area, in which the first processing of the eyes’ signals takes place. When there is no input, in case of closed eyes, this area goes to a resting state: the rhythmic pattern. Besides the primary visual area, there are many other areas processing arousal. Think for example of the primary auditory area and the sensory area, involved in processing touch senses. In front of the sensory area, which is located as a strip over the head from ear to ear, the motor area is located. From this area all voluntary muscles are controlled. The frontal area of the brain performs higher functions, like concentration, memory and emotion. Figure 15 depicts all areas of the brain and their functions.

Figure 15, some motor, sensory and association areas of the cerebral cortex.


6.5 Known frequency bands

As already mentioned the German psychologist Hans Berger found a very clear rhythm of approximately 10 Hz (or somewhere between 8-13 Hz) at the back of the head when the subject closed his eyes. He named this rhythm after the first letter of the Greek alphabet: alpha. The alpha rhythm is associated with a state of relaxation. It is also known that the alpha rhythm of infants is very low, increases till adolescence and decreases during adulthood.
Later more frequency bands were determined: beta, delta and theta. They are ordered by time of their discovery instead of frequency which makes it rather confusing. Delta is the lowest band and ranges from approximately 0.5-3 Hz. Delta waves are often seen in sleeping subjects. Theta is the next band and ranges from 4-8 Hz. This band is associated with both drowsiness and arousal and mental illness’ are often associated with the theta band power. For example children with AD/HD often show a significant excess of theta band power. The highest band is the beta band (12-30 Hz), which frequencies can be seen most prominently in the frontal areas. Beta activity is related to level of alertness. The more alert the subject is the higher the beta frequencies.

_Further reading:_ Wikipedia – Electroencephalography (sec. Normal activity)
7 The ECG frontend

As part of this assignment you have to buy and build your own ECG amplifier (ECG frontend). This electrical circuit can be used to measure the ECG of human beings. The frontend consists of a few sub circuits which will be discussed in the next subsections.

![Figure 16, impression of the ECG frontend](image)

7.1 Differential measurement principle

In section 5.6 was already mentioned that the ECG is recorded differentially, by using two measurement channels and a reference electrode. Figure 17 provides a graphical explanation of the principle that is used for the signal measurement.

The first column (A,D,G) shows how an ideal measurement looks like. Figure A and D show the voltages that can be measured on two locations on the body. The graph shows that the ECG activity is approximately opposite at two extremities (fig 17A and 17D). The quality of the measured ECG signal can therefore be improved by subtracting the signals from the two opposite electrodes (fig 17G).

In the real world it is not possible to measure such ideal signals; there is always interference from mains and other noise sources. In addition, other physiological processes in the body cause differences in skin potentials that decrease the quality of the measurements. Two strategies will be used to remove these disturbances and improve the quality of the measurement: common mode rejection and filtering:

- Disturbances caused by noise sources such as mains will be almost equal on both electrodes (Fig 17b, Fig 17E). Therefore this noise can be removed by rejecting the (noise) component that is detected on both electrodes. Fig 17H shows the effect of subtracting the two noise signals: the resulting signal is (almost) zero.

  Figure 17C and 17F show signals that will be found in practice: a weak ECG signal, inhibited with a lot of interference. After applying common mode rejection (Fig 17I) an almost perfect ECG signal is reconstructed. This signal is nearly identical to the ideal signal in Fig 17G.
Common mode rejection removes noise signals that are present on both electrodes locations with a similar amplitude. Noise components that remain in the signal, can be removed further by using filtering techniques based on frequencies. Low frequencies (and DC components) are removed by a high-pass filter. High frequencies are rejected by a low-pass filter.

Examples of real ECG measurements can be found in the appendix.

![Diagram](https://example.com/diagram.png)

**Figure 17, differential measurement principle.**
Rows: composition of realistic ECG signal (C) based on the ideal ECG component (A) and a noise component (B)
Columns show the effect of subtracting signal for: ideal ECG measurement (ADG), noise components (BEH), real ECG measurement (CFI).

### 7.2 Safety issues

Due to the electric connections to the human body the circuit has to meet very strict safety requirements. **It is therefore not allowed to connect any other circuitry to a human body other than this one.** The amplifier side of the circuit, the left side of the board according to figure 14, should always be powered by a battery and there may not be any connection from the left side of the board to the right side.

To fulfill the safety requirements the circuit is designed such that the current flowing through the electrodes under normal operation will never exceed 100 µA. When a single defect occurs, for example the operational amplifier creates a short circuit with one of the battery poles; the current will not exceed 500 µA. This kind of electrical equipment is characterized as ‘Type BF’. The B specifies the amount of current under both conditions and the F stands for floating, which means the circuit creates no connection from patient to the ground. Most power supplies realize a connection to the ground; therefore a battery is the only possible solution to power the circuit.

*Further reading:* Livenson A. R. Leakage currents in medical electrical devices ([Full text PDF](https://example.com/pdf), Only available on TU/e domain or through library proxy)
7.3 Functional description of the system

Figure 18 shows a schematic overview of the biofeedback amplifier.

Figure 18, schematic overview of biofeedback amplifier

The ECG signal is first amplified by an instrumentation amplifier. The following steps are used:

- **Remove low frequencies and DC voltages** that are not relevant for the ECG measurement. These potentials are caused by DC voltages that arise on the body due to other physiological processes. *Components: low pass filter, R1/C1, R3/C2*

- **Increase input impedance and pre-amplify signals.** By increasing the input impedance, the weak ECG signal can be measured more accurately since less current is flowing from the body into the amplifier. The signal is also pre-amplified. *Components: gain, IC1A, IC1B.*

- **Subtract signals.** A differential amplifier amplifies subtracts the two input signals and amplifies the difference between the two signals. The gain is frequency dependent so specific frequencies that are most relevant for the ECG can be attenuated. *Components: gain, IC1C/C3/etc.*

- **Common mode rejection** can be adjusted by P1.

The weak ECG signals are now pre-amplified and common mode components have been filtered. Next, the **frequency components that are not relevant for the ECG signal are removed:**

- **DC components are removed.** *Components: block DC: C4/R12*
- **High frequency (noise) components are removed.** *Component: low pass: IC1D/C5/etc.*

After filtering, a clear ECG signal remains. However, this signal is to weak to be fed into the optocoupler that has a very low input impedance. Therefore a **current amplifier** is used to increase the current of the ECG signal that is coming from IC1D to an amplitude of 80mA.

The output signal from the optocoupler is rather weak and has some artefacts. Therefore the signal is amplified *(IC3)* and high frequency noise is filtered (low pass: R23/C13). We now have a clear ECG measurement that is optically isolated from the body.

For some applications, such as feeding the signal into a microcontroller with a measurement range of 0-5V, it is convenient to shift the base level of the signal. By using **DC uncoupling,** the base level can be shifted to a preferred voltage. *(Note: this can be compared with vertical signal shifts on an oscilloscope).*

In the remaining sections, the components are explained in more detail.
7.4 Differential amplifier

The first part of the frontend (figure 19) mainly consists of a differential amplifier, also known as an instrumentation amplifier, which is preceded by some input filtering as well as an active output filter. The input filter consists of a few resistors and two capacitors. Very high frequencies, which are not expected in the ECG signal, are suppressed by this circuit. Besides the filtering the resistors also prevent the current to exceed 500 µA after a single defect, as described in the previous section.

Secondly the instrumentation amplifier amplifies the difference between both inputs, while suppressing common signals. This amount is expressed as the common mode rejection ratio (CMMR). Note that this circuit has one little difference. Opamp’s IC1A and IC1B are connected not only by a resistor, but with a resistor and capacitor in series connection (R9 and C3). This causes the amplifier to amplify less for frequencies below approximately 16 Hz compared to frequencies higher than 16 Hz. The potentiometer in the circuit can be used to improve the CMMR, but in most cases placing the slider in the middle position is already good enough. A detailed description of this subcircuit is provided by Niedermeyer et.al. (Niedermeyer, E. & Silva, F. L. d. Electroencephalography: Basic Principles, Clinical Applications, and Related Fields Lippincott Williams & Wilkins, 1999).

Further reading: Wikipedia – Instrumentation amplifier

The last part of this circuit is the filter built around IC1D. First a capacitor and resistor (C4 and R12) block the DC value in the ECG signal. The filter used is a lowpass filter, which suppresses all frequencies higher than approximately 100 Hz.

Some numbers and calculations:

- **R2:** Input impedance: $1\text{M}\Omega$
- **Low-pass filter R1/C1:**
  
  Cutoff frequency: $f_c = \frac{1}{2\pi RC} = \frac{1}{2\pi 10k\Omega 27pF} = 590 \text{ kHz}$
  
  Effect: only frequencies below 590 kHz can pass
- **IC1A / IC1B:** non-inverting amplifier with some frequency scaling
  
  $$\text{Gain} = \frac{R_6}{\text{Resistor_to_ground}}$$
Problem: in the circuit there is no resistor to the ground. However, there is a “virtual ground” is created in the circuit R9 and C3. The (complex) resistance that is experienced by the opamp is a combination of the ohm-resistance of R9 and a frequency-dependent load of C3. This means that the circuit around IC1A and IC1B don’t behave like a traditional non-inverting amplifier. The “virtual ground” is located in the circuit R9 and C3.

It goes beyond the scope of this reader to calculate the exact frequency-dependent gain of this circuit. However we can make an estimation of the gain; the “resistance to the ground” will be in the order of tens of kiloOhms ($10^4$). Since the order of R6 is also $10^4$ the gain will be of order 1.

• Differential amplifier around IC1C/R5/R7/R10/R11:

$$Gain = \frac{(R5+R10)\,(R11+P1)}{(R11+P1+R7)\,R10}$$

Potmeter P1 is used to finetune the value of R11. In practice the value of (R11+P1) is nearly identical to the value of R10.

Furthermore the circuit is symmetric: $R5 = R7 = R_A$ and $R10 \approx (R11 + P1) = R_B$

$$Gain = \frac{(R5+R10)\,(R11+P1)}{(R11+P1+R7)\,R10} \approx \frac{(R_A+R_B)\,R_B}{(R_B+R_A)\,R_A} = \frac{R_B}{R_A} = \frac{47k\Omega}{10k\Omega} = 4,7$$

Effect: signal is amplified by a factor 5

• High-pass filter C4/R12:

Cutoff frequency: $f_C = \frac{1}{2\pi RC} = \frac{1}{2\pi \times 3,3M\Omega \times 1\mu F} = 0,048 \, Hz$

Effect: only frequencies above 0,05 Hz can pass (DC block)

• IC1D: gain and filtering

7.5 Galvanic Isolation

One of the most important functions the circuit realizes is a galvanic isolation between the human body and a further processing of the signals. How this is realized is depicted in figure 19. Without this isolation a malfunctioning circuit placed behind the frontend could cause a current flow through the human body which exceeds the safety requirements.
The isolation is realized with an optocoupler. This IC (IC4) uses light to transfer the signal from one side to the other. Inside of the IC there is a LED and two equal photo diodes. Note that an ideal opamp causes its inputs to have an equal voltage, as long as there is a negative feedback. In this case the negative feedback is realized by one of the photo diodes inside the optocoupler, thus the signal on the first photo diode will be equal to the input signal. Since both diodes are equal the signal on the other diode, at the other side of the isolation, will be equal to the signal as well.

Further reading: IL300 datasheet (Full text PDF)

The opamp’s used are so called rail-to-rail opamp’s, which means that their output voltage can range up to their supply voltage. Besides this they are also capable of supplying a rather large current, needed to drive for example the LED in the optocoupler (IC2). These properties make them also very useful to be used as output driver (IC3). The output driver only copies the voltage over the photo diode to the output. The last components in this part of the circuit (R23, C13 realize again a low pass filter. At this point we have a usable signal, between −V and +V, that is filtered and can be fed into your own application.

However most microcontrollers require an input voltage between 0 and 5V, which means that the signal needs to be scaled to another voltage range. Further, it would be convenient use the same power source for the microcontroller and the isolated part of the circuit. The combination of C12 and P2 uncouples the signal and shift the signal to a voltage between V− (that can be used as ground for the microcontroller) and +5 (note: +5 volts above the microcontroller ground, not +5 volts above the original ground signal of the circuit).

The un-shifted signal is still available at test point TP2.

### 7.6 Power Sources

Because of the galvanic isolation between the input and the output of the amplifier, two separate power supplies are required for the circuit.
The opamps in input-side the circuit require both a positive and a negative supply voltage. To use a simple 9V block battery as source, a virtual ground is created by using the circuit around R18,R19,C7,C8. This results in a symmetrical +4.5V / -4.5 Voltage around the (virtual) ground. A similar circuit is created for the output-side of the circuit. Here also a +5Voltage is created that can be used to supply a microcontroller. Note: this 5 Voltage is refered to the ground of the battery (V-), not the virtual ground of the circuit itself!

7.7 Connecting the frontend, recording your own ECG

The frontend has a rather large number of connections. Both sides of the PCB are provided with gray screw terminals. Connectors to the amplifier circuit are placed on the left side, according to figures 14 and 18. A power connector is placed in the middle. As mentioned before: this power supply always has to be a 9 volt battery! The connectors at the top and bottom left are the signal inputs together with ground connectors to shield the electrode wires. An overview to the connections is provided in figure 21.

Figure 21, connections on the ECG frontend

All output connections can be found at the right side of the PCB. The signal output is located at the bottom and the microcontroller outputs are located at the top. The power connector is again located in the middle. This part can either be supplied by a 9 volt battery or power supply set to 9 volt.
8 Your design idea

During the assignment an overview of the sources and methods to measure bio-electricity has been given. It is now up to you to think of a useful, inventive, attractive, etc. application in which you use the recorded ECG signal.

If you are still not sure which information, provided by the signal, to use; think for example of the heart rate as indication of effort or reflecting the state of your subject. An increase of heart rate combined with sweating, which decreases the skin resistance, could indicate somebody is lying. Another measure is the heart rate variability, which information can be extracted from this measure can probably be found on the internet. By using the raw signal one could also detect some major heart faults, but this might be difficult already and cannot be demonstrated easily.

Make couples to work on this final task. Send in your design idea by email three days before the fourth meeting. Prepare a short presentation of your idea during the fourth meeting. Use for example sheets of drawing paper or a simple electronic presentation. After each presentation we will discuss the idea and there will be some questions for each couple to be answered.

The fifth meeting is used for answering questions and to provide some feedback on your brilliant ideas. Within a week after this meeting each couple has write a small report which describes their idea and the results obtained.
A. Microcontroller DSP firmware

This appendix will discuss a possible implementation of a digital signal processing system to detect the QRS complex in the ECG signal that can be implemented as firmware in a microcontroller. First a simple circuit is introduced. Later the firmware is discussed in more detail. To be able to understand these DSP operations a (extreme) short introduction to DSP is given. Note that the concepts discussed in this single chapter are usually spread over a few years and different courses for an electrical or mechanical engineering student.

Note: Beside the implementation discussed in this section, also an DSP implementation for the ARduino is available. For the most recent version see our website (www.biofb.nl)

8.1 Microcontroller circuit

The circuit provides a well conditioned ECG signal directly usable for many applications; however the circuit also provides an analog voltage, which corresponds to the current heart rate, and a short pulse during every heart beat. These signals are generated by using a microcontroller which uses digital signal processing (DSP). Besides a microcontroller a few passive components are used to realize the circuit, which is depicted in figure 20.

![Figure 22, digital signal processing part of the ECG frontend](image)

The microcontroller detects the QR flank of the QRS complex (how this is done is explained in the appendix) and counts the number of timer overflows between each QR. This value is used to set the PWM value. The human heart rate can vary between 60 and 200 BPM which will be converted to an analog value of approximately 0 to 4.2 volt, therefore shifting and scaling is needed. Once the PWM value is determined the PWM unit inside the microcontroller operates autonomously at a frequency of 37.5 KHz. This frequency can easily be lowpass filtered by R27 and C12.

To create the pulse, the pulse output is set to a high value when a QR flank is detected. Analog the RS flank is detected which toggles the pulse output to a low value. This output is provided with a LED to visualize the operation of the microcontroller.

Further reading: ATTINY13 datasheet (Full text PDF, Chapters ‘Analog to Digital Converter’ and ‘8-bit Timer/Counter0 with PWM’ are relevant)
8.2 Digital Signal Processing

First consider the basic concept of signal representation: every periodic signal of length $1/f$ can be represented as a (infinite) summation of sinusoids with frequency $n \cdot f$, with $n = 0, 1, 2,...$ A square wave for example can be approximated as:

$$f(t) = \lim_{N \to \infty} \frac{4}{\pi} \sum_{n=1}^{N} \frac{1}{2n-1} \sin\left(2\pi \cdot (2n-1) \cdot t\right)$$

This can easily be verified by putting this formula for $N = 5$ into your graphical calculator. Of course this concept also holds for our (periodic) ECG as we will see later.

Digital signal processing starts by sampling the signal we would like the process, in this case an ECG signal. Sampling is nothing more than measuring the voltage level at a fixed rate, in our example 200 times a second. A simple example of a sampled signal is given in figure 22. This example is a 50 Hz sinusoid. As long as the frequencies in the signal (think of the infinite sum) are less than half the sampling frequency, the original signal can be completely reconstructed form the sampled values. The latter is known at the Nyquist theorem.

![Figure 23, sampling of a periodic signal](image)

A 250 Hz sinusoid, also sampled at 200 Hz, will pass through exactly the same values, but will be reconstructed as a 50 Hz sinusoid; this is known as aliasing, which is not favored of course. This example indicates the importance of choosing the right sample frequency. Because the microcontroller onboard of the frontend is not able to process thousands of samples every second a low sampling frequency of 200 Hz is chosen, which is no problem since there are not much frequencies higher than 100 Hz in the signal. As will be discussed later this sampling frequency is also chosen for another reason.

Let’s consider the ECG signal again, which is depicted in figure 23 (sampled at a much higher rate than necessary to capture all interferences as well). It is very clear that the signal still contains a lot of mains interference. When this signal is sampled in the microcontroller, the voltage level can only be stored as an integer number. The input voltage is converted to a number between 0 and 255 (the maximum number which can be stored in 8-bits) according to which fraction of the reference voltage it is.

$$x = \left\lfloor \frac{V_n}{V_{ref}} \cdot 256 \right\rfloor$$
Although a 10-bit ADC is included in the microcontroller its last 2 bits are not used, since we do not need such precision. The sampled signal is depicted in figure 26.

![Figure 24, example of a recorded ECG](image)

To obtain a good detection of the QRS-complex in the signal, the mains interference will have to be suppressed first. This could be done by adding an analog filter before digitizing the signal. However we do not know the exact frequencies in the signal and the filter should be adjustable, without changing electric components. The only way to fulfill both requirements is to implement the filter digital.

There is one thing known about the signal and its interference: the interference is approximately 50 Hz (and maybe some 100 Hz; light and electromagnetic fields of TL-lamps have this frequency), while the signal is mostly spread over other frequencies. Therefore the filter should suppress frequencies at 50 and 100 Hz. At this point it will be obvious why 200 Hz was chosen as sample frequency. A signal of 50 Hz is repeated every 4 samples (and a 100 Hz signal every 2), so if 4 succeeding samples are added the contribution of the 50 (and 100Hz) part will be zero (see figure 25). If one adds the second last sample to the current only the 50 Hz part will be suppressed. The latter is known as a ‘notch’ filter since it only suppresses a specific frequency, while the former could be named ‘smoothing’ filter, but this is not an official name. The smoothing filter will be discussed further, since it yielded the best results.

![Figure 25, smoothing filter](image)

To implement the filter one will have to remember the current, last, second last and third last sample, add them and scale to make sure it can be stored in 8 bits. These operations can be implemented on a microcontroller very well. As an equation y of terms of x can be described as:

\[ y[n] = \frac{x[n] + x[n-1] + x[n-2] + x[n-3]}{4} \leftrightarrow Y(z) = \left(1 + z + z^2 + z^3\right) X(z) \]
At the right side of the arrow the Z-transform of the transfer is given. More information about Z-transforms can be found at for example Wikipedia (Z-transform). A Z-transformed transfer, simply called transfer function, shows how the output of a filter is related to the input and in particular the frequencies in both signals. Which $z$ is used depends on the sample frequencies of the system. In general

$$z = e^{j2\pi f/T}$$  \hspace{1cm} (1.1)

is used, which is a complex number on the unit circle (circle described by

$$\sqrt{\text{Re}(x)^2 + \text{Im}(x)^2} = 1$$, on the complex plane, horizontal real part and vertical imaginary part of the number, see figure 24).

Further reading: Wikipedia – Complex number

The transfer function of the filter

$$H(z) = \frac{Y(z)}{X(z)} = 1 + z + z^2 + z^3$$

is equal to 0 for $z = i, -1, -i$, which correspond to $f = 50, 100, -50$ Hz according to equation 1.1 (as long as $f_i = 200$Hz of course). The ‘zeros’, as these values of the transfer function are called, are depicted in figure 25 as small circles. The transfer magnitude of all frequencies, depicted in figure 24 can, be derived by ‘walking’ over the upper half of the unit circle form 1 to -1. At and close to the zeros the signal is suppressed, which result in a negative magnitude in dB.

![Figure 26, zero/pole plane for the smoothing filter](image)

It is understandable if the last few paragraphs were a little confusing for somebody with no engineering background. Unfortunately it is not possible to explain all properties of signal representations, signal transforms and transfer functions. However there are many pages with excellent information available on the internet.

The results of the filters can be depicted in Matlab. As mentioned before a notch as well as a smoothing filter was used. The results obtained with the smoothing filter were better than those of the notch filter, which can be seen in figure 26. The notch filtered signal still fluctuates a lot, while the smoothed signal almost flat except for the PQR-complex. It is now even possible to recognize the P top (at sample 6), something not possible in the original signal.

![Image](image1.jpg)

**Figure 27, sampled and quantized (filtered) signals**

The last thing to do is determine when the PQR-complex begins (and ends) in the signal, to be able to turn the LED on and off. Of course this could be implemented by comparing with a certain threshold, but this demands that the signal always has the same shape and mean. A better way is to differentiate, which is quite simple with a discrete signal, and then compare it with a high and low threshold. This way strong increments and decrements are detected. The differentiation is implemented as a subtraction, in which the differentiated signal is the current sample minus the previous sample. In this example the differentiated version of the smoothed signal is depicted in figure 27. The low and high thresholds are 15 and -10 respectively. The lower threshold is chosen rather low since we would always like to turn off the LED.

![Image](image2.jpg)

**Figure 28, differentiated smoothed signal**

Every time the beginning of the QRS complex is detected the microcontroller updates the PWM register to generate a new DC voltage. When a QRS-complex is missed, because the differentiated signal is not high enough, the time between the previous complexes is used, which means the PWM register is unchanged. The microcontroller knows that a complex is missed when the time between complexes changes too rapidly.
B. Building description

This document briefly describes how to build the ECG frontend belonging to the assignment “Electrical measurements on the human body”.

On the next page of this document there is a list of all components included in the package. Start by checking if all components are actually included. Missing resistors or capacitors can be obtained at the … If there is some other part missing please contact … Besides amount and value, this table includes a color code column for the resistors.

Building the frontend is just a matter of putting all components in the board (figure 2), but there is a preferred order to do this:

1. Put the resistors in the board and solder them. Again make sure to pull them straight on the board.
2. Insert the IC-sOCKETS. Mind the small notch on one side, which has to be on the same side as in the figure. To keep the sockets in place bend two pins at opposite sides, while you solder the others. Now bend back the ones that are bended and solder them.
3. Insert the potentiometers. Again fix it in the board by bending the wires just a little apart from each other.
4. Insert all capacitors (common as well as electrolytes). Mind the positive side of the electrolytes which is indicated as a small plus sign in the figure.
5. Put in the wire screw connectors on both sides of the board.
6. Insert all IC’s in the IC-sockets. Again mind the little notch on both socket and IC package. The notches have to be on the same side (at least as long as you have put the sockets in as described above). If there is no notch on the IC, then there is a small hole which marks pin 1. This is at the same side as the notch.
7. Fix the provided legs under the board by screwing them in the holes.
8. Connect batteries and leads as shown below.

Connections on the ECG frontend
<table>
<thead>
<tr>
<th>Type</th>
<th>Value</th>
<th>Amount: ComponentID</th>
<th>Color code</th>
</tr>
</thead>
<tbody>
<tr>
<td>Resistor</td>
<td>100</td>
<td>2 : R23, R24</td>
<td></td>
</tr>
<tr>
<td></td>
<td>220</td>
<td>1 : R17</td>
<td></td>
</tr>
<tr>
<td></td>
<td>4k7</td>
<td>1 : R15</td>
<td></td>
</tr>
<tr>
<td></td>
<td>10k</td>
<td>9 : R1, R3, R5, R7, R9, R18, R19, R20, R21</td>
<td></td>
</tr>
<tr>
<td></td>
<td>22k</td>
<td>2 : R6, R8</td>
<td></td>
</tr>
<tr>
<td></td>
<td>43k</td>
<td>1 : R11</td>
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<td>47k</td>
<td>3 : R10, R16, R22</td>
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<tr>
<td></td>
<td>150k</td>
<td>1 : R14</td>
<td></td>
</tr>
<tr>
<td></td>
<td>1M</td>
<td>2 : R2, R4</td>
<td></td>
</tr>
<tr>
<td></td>
<td>3M3</td>
<td>2 : R12, R13</td>
<td></td>
</tr>
<tr>
<td>Potentiometer</td>
<td>10k (multiturn)</td>
<td>1 : P1</td>
<td></td>
</tr>
<tr>
<td></td>
<td>500k (vertical)</td>
<td>1 : P2</td>
<td></td>
</tr>
<tr>
<td>Capacitor</td>
<td>1n</td>
<td>1 : C13</td>
<td></td>
</tr>
<tr>
<td></td>
<td>10n</td>
<td>1 : C5</td>
<td></td>
</tr>
<tr>
<td></td>
<td>22p</td>
<td>1 : C6</td>
<td></td>
</tr>
<tr>
<td></td>
<td>27p</td>
<td>2 : C1, C2</td>
<td></td>
</tr>
<tr>
<td></td>
<td>100n</td>
<td>1 : C11</td>
<td></td>
</tr>
<tr>
<td></td>
<td>1u</td>
<td>3 : C3, C4, C12</td>
<td></td>
</tr>
<tr>
<td></td>
<td>100u (elco)</td>
<td>4 : C7, C8, C9, C10</td>
<td></td>
</tr>
<tr>
<td>Opamp</td>
<td>TL071</td>
<td>2 : IC2*, IC3</td>
<td></td>
</tr>
<tr>
<td></td>
<td>TL074</td>
<td>1 : IC1</td>
<td></td>
</tr>
<tr>
<td>Linear optocoupler</td>
<td>IL300</td>
<td>1 : IC4</td>
<td></td>
</tr>
<tr>
<td>Voltage regulator</td>
<td>78L05</td>
<td>1 : IC6</td>
<td></td>
</tr>
<tr>
<td>IC-socket</td>
<td>DIL8</td>
<td>3 : IC2, IC3, IC4</td>
<td></td>
</tr>
<tr>
<td></td>
<td>DIL14</td>
<td>1 : IC1</td>
<td></td>
</tr>
<tr>
<td>Screw terminal</td>
<td>2 pin</td>
<td>2 : X1, X3</td>
<td></td>
</tr>
<tr>
<td></td>
<td>3 pin</td>
<td>2 : X2, X4</td>
<td></td>
</tr>
<tr>
<td>Board</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Bill of materials

*Note: For IC2 also a TS921 rail-to-rail opamp can be used. This opamp can provide a higher output current of 80mA (TL071: 40-60mA)

TIP: When measuring EEG or EMG signals the gain can be increased a factor 10 by increasing R10 and R11 a factor 10 (470k)
PCB Top layer (view from top)

PCB Bottom layer (view from top)
C. Specifications

The specifications for the default biofeedback amplifier (ECG mode) are listed below. Amplifiers that are optimized for ECG, EMG, GSR or EOG measurements have different specifications.

<table>
<thead>
<tr>
<th>Specification</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Input impedance</td>
<td>&gt; 1MΩ</td>
</tr>
<tr>
<td>Dynamic input range</td>
<td>5mV&lt;sub&gt;u&lt;/sub&gt;</td>
</tr>
<tr>
<td>Supply current measurement amplifier</td>
<td>11mA</td>
</tr>
<tr>
<td>Supply current optocoupler</td>
<td>2.2mA</td>
</tr>
<tr>
<td>Common mode rejection (CMRR)</td>
<td>&gt; 70dB</td>
</tr>
<tr>
<td>Amplifier gain</td>
<td>60dB (1000x)</td>
</tr>
<tr>
<td>Bandwidth (depending on filter config)</td>
<td>0.4 … 35Hz</td>
</tr>
<tr>
<td>Supply voltage</td>
<td>9V battery</td>
</tr>
<tr>
<td>Output voltage microcontroller port</td>
<td>5V</td>
</tr>
<tr>
<td>Output current microcontroller port</td>
<td>85mA</td>
</tr>
<tr>
<td>(can be increased by replacing 78L05 with a more powerful current regulator)</td>
<td></td>
</tr>
<tr>
<td>Output signal microcontroller port</td>
<td>0-5V</td>
</tr>
<tr>
<td>(level shift available)</td>
<td></td>
</tr>
</tbody>
</table>

Connections on the ECG frontend

X2:
- Lead 1
- Ground Lead
- Lead 2

X1:
- Battery –
- Battery +

X3:
- Battery –
- Battery +

X4:
- +5V Out
- Signal Out
- GND