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FACULTEIT INDUSTRIËLE INGENIEURSWETENSCHAPPEN
master in de industriële wetenschappen: elektronica-ICT

Masterproef
EEG signaal analyse

Promotor :
Prof. dr. ir. Bart VANRUMSTE

Promotor :
prof. dr. ir. JOS KLAPS

Gezamenlijke opleiding Universiteit Hasselt en KU Leuven

Frederik Vreys
Scriptie ingediend tot het behalen van de graad van master in de industriële wetenschappen: elektronica-ICT

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Preface

This project started with the request to improve the time resolution precision of an existing measurement system for sleep-EEG(electroencephalogram) signals. Not only does this lead to very precise showings of the EEG signals but more importantly to software defined trigger points.
It turned out that improving the existing system was impossible due to encrypted signals and since the manufacturer did not share the decryption the improvement lead to an end. Instead this project became the start of the development of a new sleep-EEG measuring system one which fulfils al the demands and is the first in its kind of EEG measuring systems usable at home while still a very accurate for diagnosis.

At the first stage of this project a simulation program and hardware setup was built in order to prove that the concept was possible. However, putting this concept into a product felt like the real challenge provided that the sensor needed to be wireless and that the software setup and part of the hardware setup needed to be embedded into the sensor.
Knowing that this project is just in its infancy and that it comprises software in multiple languages and hardware development, following it up in a master thesis when given the opportunity to do so expands the field of knowledge in more than one direction.

The project was started by Dr. ir. J. Klaps. Through his knowledge of sleep-EEG patterns, diagnosis and studies he devised the goals for this project. The methods of J. Klaps's acquaintances in the field of brain research served as the blueprint for the properties of the EEG measurement system. Furthermore, his support in lab material made the production and testing of the EEG-sensor optimal and efficient.
Prof. Dr. ir. R. Thoelen was assigned by FIIW (Faculteit Industriële Ingenieurs Wetenschappen) as the internal promotor. His knowledge and experience of data acquisition in microcontrollers and LabView directed the design into the right directions. Because he shared his resources, e.g. for making device cases, the project developed quickly.
Louis Lenearts is thanked for loaning his material and practical knowledge.

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List of Abbreviations

ADC	Analogue-to-digital converter
AP	Action potential
ASCII	American standard code for information interchange
CMR	Common-mode rejection
CNS	Central nervous system
DC	Direct current
ECG	Electrocardiography
EEG	Electroencephalography
EMG	Electromyography
ESD	Electrostatic discharge
G	Gain
GND	Ground
GUI	Graphical user interface
I^2C	Inter-integrated circuit
IC	Integrated circuit
INA	Instrumentation amplifier
IPSP	Inhibitory postsynaptic potential
LED	Light emitting diode
LiPo	Lithium polymer
MAC	Media access control
MXP	myRIO expansion port
NI	National Instruments
PCB	Printed circuit board
RC	Resistor-capacitor
RLD	Right leg drive
Rx1	Receive pin
SVE	Shared variable engine
Tx1	Transmission pin
UART	Universal asynchronous receiver/transmitter
USB	Universal serial bus
UVLO	Under voltage lock out

List of Symbols

µF	Microfarad
µV	Microvolt
µV/°C	Microvolt per degree celsius
B	Precedes a word of bits
C_x	Capacitor
dB	Decibel
F_{p1}	Left front electrode of the sensor
F_{p2}	Right front electrode of the sensor
F_{pz}	Middle front electrode of the sensor
G	Gain
Hz	Hertz
K	Amplifying factor
kΩ	Kiloohm
mA	Milliampere
mAh	Milliamperehour
mm	Millimeter
mV	Millivolt
Q	Quality factor
R_G	Gain-resistor value
R_x	Resistor
S/s	Samples per second
V	Volt
Vcc	Power supply (used interchangeably with Vdd)
Vdd	Power supply (used interchangeably with Vdd)
Ω	Ohm
Ωm	Resistivity (simplified form, full form: $\rho=\Omega*m^2/m$)

Abstract

The mission of this project is to build a system with a high time resolution to investigate the common neurophysiological mechanism of abnormal brain functioning clustering groups of pathologies with EEG-recorded parameters in conjunction with subjective parameters which are acquired after software defined forced awakes during the sleepcycli.

Most sleep-EEG measurement systems at home are not built for diagnostics but mainly for personal interests, hence a low time resolution for these systems is sufficient. However, having a low time resolution makes these systems far from ideal as medical diagnostic sensors.

The unavailability of diagnostic home measurement systems lead to the design of a wireless EEG-sensor which transmits the measured signals towards the real-time device 'NI myRIO' from National Instruments where the signals are processed. The EEG-sensor exists out of a microcontroller, amplifier, low pass filter and RN-42 Bluetooth module which also resides on the 'NI myRIO'. The microcontroller samples synchronously the amplified and filtered EEG signals and transmits them immediately. In LabView the signals are filtered into different kinds of brainwaves and processed. If certain conditions of the brainwaves are met they will be acted upon.

This EEG measurement system is, not only because of its high time resolution and conditional actions, an excellent medical sensor. It also creates many opportunities for future applications such as EEG processing smartphone applications thanks to its flexible software.

Abstract in het Nederlands

De doelstelling van dit project is een systeem met een hoge tijdschaalresolutie te bouwen om zo het neurofysiologische mechanisme te analyseren van abnormale brein functionerende clustering groepen en de pathologieën met EEG-metingen en subjectieve parameters, bij een geforceerd ontwaak, tijdens de slaapcycli.

Bestaande EEG-meetsystemen voor thuisgebruik zijn niet gebouwd voor medisch onderzoek maar voor persoonlijke interesse en hebben daardoor voldoende aan een lage tijdschaalresolutie. Dit is echter niet voldoende voor medisch onderzoek.

Het ontbreken van EEG-meetsystemen voor medisch onderzoek leidde tot het ontwerp van een draadloze EEG-sensor die de gemeten signalen doorstuurt naar 'NI MyRIO', een real-time toestel van National Instruments, waar de signalen worden verwerkt. De EEG-sensor bestaat uit een microprocessor, een versterker, een laagdoorlaat filter en een RN-42 Bluetooth module. Op 'NI MyRIO' bevindt zich dezelfde Bluetooth module. De microprocessor samplet de versterkte en gefilterde signalen synchroon en verzendt ze. Vervolgens worden de signalen in LabView gefilterd in verschillende soorten hersensignalen en verwerkt, zodat er op gereageerd kan worden indien de hersensignalen aan een bepaalde toestand voldoen.

Dit EEG-systeem is, niet alleen door zijn hoge tijdsresolutie en mogelijkheid tot interactie op specifieke hersenactiviteit een volwaardig medisch systeem. Het creëert ook ruimte voor toekomstige smartphone applicaties omwille van de flexibele software.

1 Introduction

This chapter starts by discussing the vindication of this project by giving background information. The chapter thereafter describes the problem succinct. This problem definition is then translated into practical objectives in the following section. The methods and materials that are used to accomplish these objectives are summarized in the fourth section. Finally, a short outline of the thesis is provided.

1.1 Background

Everybody who listens to the news has heard it, top sports people failing in their activity. The conclusion is always the same, the cause could have been prevented if a better medical trail had been made. However, top sports people nowadays have received descent medical trails before the starting of their official career. This leads to the conclusion that one medical trail is far from enough meaning that the results from the trail have been false-positive. There is only one solution and it is the prolonged supervision only then you can be sure that you will once measure the abnormality. For cardiac supervision the Insertible Cardiac Monitor[1] is used. It measures the ECG signals (electrocardiography).

The same story accounts for sleep-disorders. For a prolonged period of time the medical science has used EEG signals as a diagnostic tool for patients with sleep disorders. However, today patients still have to sleep in the hospital while being watched and logically speaking you cannot have a patient sleep for more than a week in the infirmary. If the abnormality would not happen or if the different sleep environment and supervision feeling takes the overhand during that night sleep this could lead to false-positive results or even inconclusive results. The physician would then probably give the wrong medication and usually they just prescribe the usual. This renders the trial at the hospital useless altogether.

Sleep examinations take place in a sleep laboratory at an infirmary and doctors need to be permanently present. The main reason for this is the operation of the device, it needs trained personal because it is a difficult apparatus. But in the end the measuring apparatus used in laboratories produce very precise results. Subjective parameters are also denoted and acquired through the use of an intentional awakening and subsequently a question which is most of the time of a rating from 0-4.

Personal interest in sleep data has increased a lot over the last few years. The market responded with personal sleep monitors. Doctors' interest in these monitors is also increasing because of the monitors' ability to be used at home. Indeed, the physician does not have to be present all the time anymore.
One monitor for example is the myZEO(Figure 1) and although its measurements and processing is very accurate its time basis precision, like many other monitors, is very low.

Figure 1: Personal sleep monitor MyZEO

MyZEO in particular has a minimum time basis unit of 5 minutes which is enough to differentiate between different sleep phases and complete cycli. However it is far too ample for medical diagnostics, because the important EEG signals take place at the transition between sleep phases. Nevertheless personal monitors have an advantage over sleep laboratories in being usable at home. In case of a sleep laboratory the patient is going to be examined in a different bed, environment and under nonstop supervision. It is very likely that the patient will exhibit a different sleep behaviour which will lead to a wrong diagnosis. Also, current laboratories measure with inconvenient leads. With a diagnostic system at home the patient does not experience these discomforts and will exhibit the normal sickly sleep behaviour and even though it does not happen that often, with prolonged supervision it is almost certain that it will be measured one day.

The EEG monitor developed in this thesis will be the first to experiment with a prolonged supervision of sleep disorders by measuring the EEG signals.

1.2 Problem Definition

A universal EEG measurement system does not exist. The fact that there are various ways of defining the spectrums for brain-waves makes certain systems ideal for some and useless for others.

Furthermore you have the examination in a sleep laboratory that contains the problem of the ‘Big Brother is watching you’ feeling of the patient. Also, these examinations are performed by using non-wireless equipment which is uneasy for the patient. The current technology also requires trained personal to operate it.
The personal sleep monitor at home which has a problem with the very low time resolution. Because, these are the only two different kind of systems available on the market a diagnosis will always experience one or more of these problems.

Thus, the ward concerned with sleep disorders is in need of a highly adjustable home EEG measurement system with the assets of both the measurement system used at a sleep laboratory and the personal sleep monitors that can be used at home.

1.3 Objectives[1]

The system should have two operable ways. One of the two is the use as a bedside system and the other is the use as a real-time system.

For starters, the bedside mode should require as few interaction from the user as possible. Ideally the user should only have to pick up the sensor and wear it during the sleep. One exception, the sleeper has to register a rating at certain important points in the sleep cycle. The sensor itself should be wireless, small and comfortable.

The real-time system still requires the patient's interaction to be minimal but this mode also ought to enable the doctor to interact with the system. The interaction includes being able to view the signals in real-time, waking up the patient on demand, changing the spectrums that should be measured and in addition be able to create settings for future days when the bedside mode will be used.

Subjective data is required when a certain state in the sleep has occurred. These states have to be detected programmatically. In addition these states should have the following parameters:

- the spectrums that need to be measured and saved;
- specific frequencies that need to be measured and saved;
- the conditions to wake a patient;
    - the amplitude of the signal,
    - the duration the signal has to be above that amplitude;
    - in which part of the night,
    - how many times the patient should be woken for the same conditions;
- the rating range.

Consequently, when it is detected the patient should be woken up.

Last but not least and certainly not least important, the data should be stored onto a USB(Universal Serial Bus) flash drive.

[1] Regarding the CE-marking, this device will be used as a research use only device. Therefore the rules to receive a CE-marking are not fulfilled. However, several components with a CE-marking are already used. Would it one day be desired to receive the marking, the CE-marked components will make the adaptations of the sensor less tedious.

1.4 Methods and Materials

This project already had the National Instruments (NI) myRIO[2] at its disposal. It is a real-time device capable of connecting to a network, but more importantly it can run stand-alone-applications and thus fulfils the bedside mode requirement. It will be used as the central processing unit and since it is a National Instruments product it should be programmed in LabView. To make the program easy and readable by people who did not program it, finite-state machines and producer-consumer loops have been used. These structures also make it more easy to add code.

The ability to connect the myRIO to a wireless network allows it to publish network shared variables[3]. A personal computer on the same network is then able to access these shared variables. They can be written to or read from. Hence they can serve to make the system interactive. Moreover shared variables are also accessible with the Data Dashboard for Labview application (AppStore, googlePLAY) from National Instruments[4]. Therefore a smart device can be used to register subjective parameters from the patient.

The sensor measures the EEG signals by means of an analogue amplifier and filter circuit. Subsequently it will be digitized by an analogue-to-digital converter (ADC). In the last step it will be wirelessly transmitted to the myRIO. The sensor contains a tiny microcontroller called Teensy 3.1[5] in order to control the ADC and transmission.

The sensor also contains a temperature sensor.

The sensor will be made to be safely rechargeable meaning that it can stay on the charging station without supervision.

The charging station will be made on the myRIO expansion port(MXP) protoboard connector. That way the myRIO can sense whether the sensor is in use or not.

Moreover, the MXP connector also exhibits a Universal Asynchronous Receiver/Transmitter (UART) port. This comes in handy for the wireless communication. Indeed, it enables us to use Bluetooth modems that can be accessed through UART. A Bluetooth module will then be connected to the myRIO and one can be connected to the microcontroller residing on the sensor which also has a built in UART port. The Bluetooth modem that is used is called the BlueSMiRF Silver.[6]

Bluetooth is the most suitable form of wireless transmission for the sensor. It does not nearly use as much power as Wi-Fi does and the connection does not rely on a network as is the case with Wi-Fi.

To make this system as flexible as possible regarding the bedside system mode and real-time mode the program will not differentiate between these two. That way on whatever point in time there can be switched between these modes.

1.5 Outline

Chapter 2
To enlighten the reader what the sensor measures and to which extent this paper starts with the secondary research that explains the biological and diagnostic background of this project. The biological events and diagnostic purposes justify the design choices and process methods which were made. Hence they should be explained thorough.

Chapter 3
This chapter focuses on the design of the sensor. It is designed so that every Bluetooth enabled device is able to connect to it and able to process the transmitted measured signals. It is made user-friendly meaning that no trained personnel is needed to operate the sensor.

Chapter 4
Chapter 4 describes the functions of the myRIO which matter the most for this project. Shared variables in particular serve a very important function. Moreover, if they did not exist the graphical user interface would not be as efficient as it is.

Chapter 5
The interface software and the myRIO software will be explained separately. Keep in mind that the program was built in such a way that it does not differentiate between bedside mode and real-time mode. Therefore interaction with the program is possible at many places, but is not necessary as the program is able to continue on its own.

Unlike most programming languages, LabView is a graphical programming language. This makes it difficult and unclear to attach the code to the thesis through the use of screenshots. Therefore a compact disc is attached to the last page which contains all the code.

Chapter 6
This chapter summarizes as much as possible what the results are a physician can expect when using the system.

Chapter 7
This chapter concludes the result of the thesis and summarizes what has been achieved. Furthermore, unsolved problems are discussesd as well as several suggestions for ongoing development of this system

2 Literature Study

Before the technology of the sensor and the technique of the processing of the signals is discussed, this chapter will cover the biological and medical aspects of EEG.
First of all the origin of EEG signals will be explained followed by the discussion of the different kind of brainwaves.
Finally the last two sections discuss the general characteristics an EEG measurement system should have.

The theory discussed is based on the first chapter of EEG signal processing by Saeid Sanei and Jonathon Chambers[7].

2.1 The origin of EEG Signals

2.1.1 Neural Activities

The central nervous system (CNS) mainly consists of nerve cells and glia cell located between neurons. The nerve cells consists of axons, dendrites and cell bodies. The activities in the CNS are related to the synaptic currents transferred between axons and dendrites or between just between two dendrites. A nerve cell typically has a negative potential of 60-70 mV in reference to extracellular environment.[7]

When a nerve cell is fired an action potential (AP) travels along a fibre. The action potential can be seen as the information transmission of a nerve cell. It is initiated in the cell body and travels in one direction. After the nerve cell fires the membrane potential depolarizes and becomes positive or at least more positive. Next, it will repolarize. It becomes more negative than the resting potential of the membrane and eventually will rise till it reaches the resting potential where it will stay till the next AP. The depolarization and repolarization take place within 1 millisecond, hence it appears as a spike. The total refactor period in which no other APs can occur is approximately 2 milliseconds. Figure 2 displays a graph of an example action potential.[7]

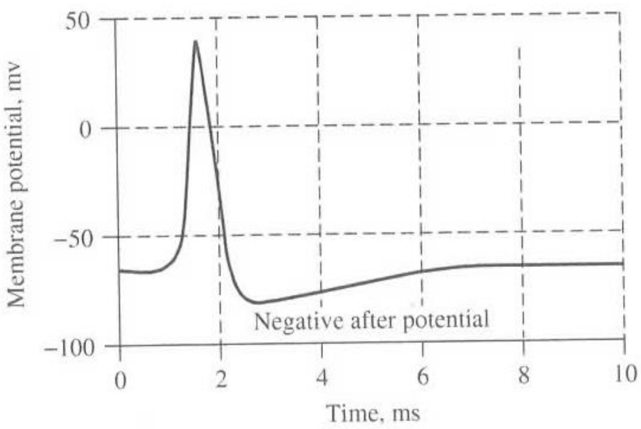

Figure 2: An action potential [7]

If the fibre of a nerve ends in an inhibitory synapse then hyperpolarisation will occur, which is called an inhibitory postsynaptic potential (IPSP). When an IPSP is generated there will be an inflow of anions or an outflow of cations across the membrane. This flow of ions, which is essentially a transmembranous current, causes the membrane potential of the nerve cell to change. Furthermore the primary transmembranous current generates secondary currents along the membrane. Sure enough, these are generated intra- and extracellular. It is the extracellular current which causes the potentials which are the beloved EEGs. The IPSP is depicted in Figure 3.[7]

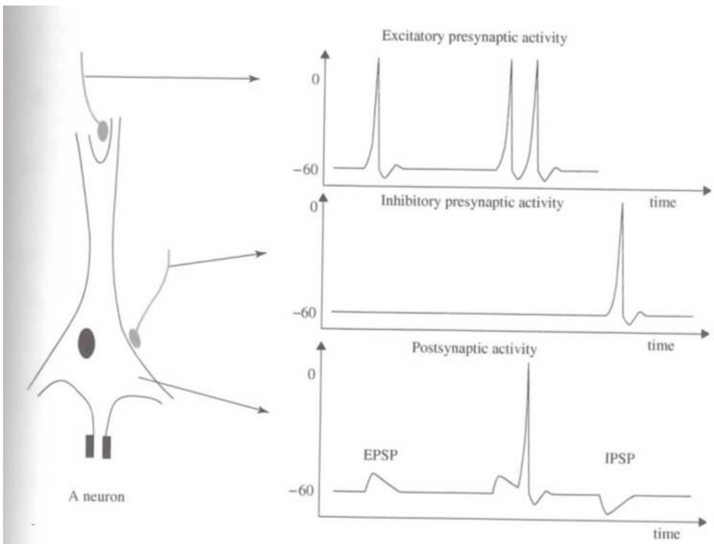

Figure 3: Nerve cell with IPSP depicted in in the bottom graph[7]

2.1.2 Measureable EEG Signals

An EEG signal is a measurement of summed secondary electrical fields caused by primal currents that flow during synaptic excitations of the dendrites located in many pyramidal neurons in the cerebral cortex.

Figure 4 illustrates the three layers of the human head with the resistivities and thicknesses of each layer. Starting from the outside the human head consists out of the scalp, skull and brain. Figure 4 also gives the resistivity of the individual layers. As can be seen, the skull attenuates about a 100 times more than the brain or scalp. However, the noise coupled onto the EEG signals are generated internally by the brain or externally by the scalp. The electrical resistivity of the scull is the main reason why measuring EEG signals is not as straight forward as it is with ECG or electromyography (EMG) of the muscles.[7]

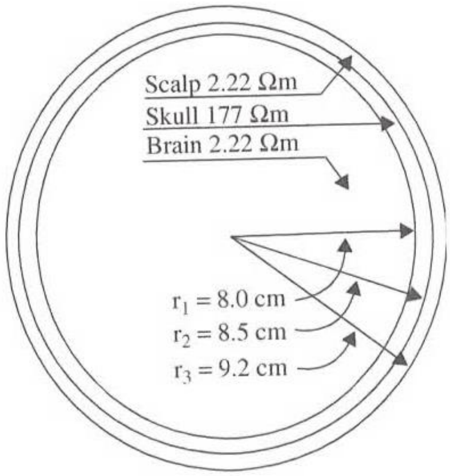

Figure 4: The three layers of the brain[7]

In conclusion, EEG signals are only significantly measured at large populations of active neurons, because it is at these locations that the summed potential of the secondary current running alongside the membrane of the nerve cell is large enough to survive till it reaches the electrodes.[7]

2.2 The Brainwaves

In total there are 5 different kind of brainwaves which can be defined by the frequency ranges. From low to high frequency ranges, Table 1 depicts them.

Table 1: The 5 different kind of brainwaves

Name	Lower frequency (Hz)	Higher frequency (Hz)
Delta(δ) waves	0.5	4
Theta(θ) waves	4	7.5
Alpha(α) waves	8	13
Beta(β) waves	14	26
Gamma(γ) waves	30	45

Figure 5 shows a measurement of 4 different kind of brainwaves. The waves are all on the same absolute amplitude scale. EEG signals start of as a summation of the potentials coming from large populations of neurons. The total potential has amplitudes up to several millivolts, but due to the layers of the human head it gets attenuated greatly to a range of 10-100 μV.[7]

Although frequency regions are listed here it does not mean that these are international standards. On the contrary, every ward has its own frequency regions which it uses to study the brainwaves.

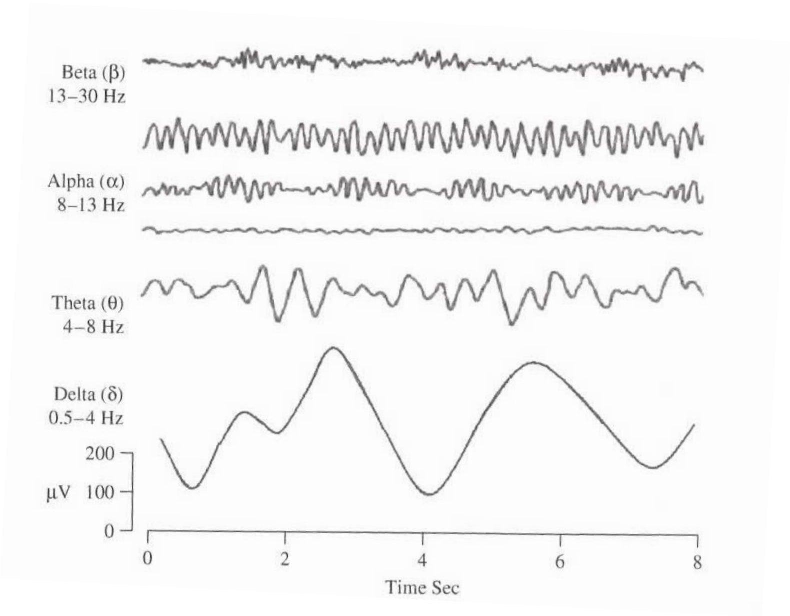

Figure 5: Example of a measurement of 4 different kind of brainwaves[7]

2.3 Conditioning and Digitizing

Raw EEG signals have a magnitude in the order of µV which is, like previously stated, the consequence of the electrical resistivity of the scull. In order to get more specific measurements the signals are amplified and filtered before they are digitized by the ADC. Most filters consist of a high pass filter and a low pass filter with cut-off frequencies of 0.5 Hz and 100Hz respectively. The high pass filter removes unwanted DC (Direct Current) and low frequency components such as breathing. Obviously, since EEG signals have a maximum frequency of 100Hz, signals beyond that frequency are considered noise and therefore filtered out of the analogue signal. What follows is the amplification of the signal. It usually starts with a differential amplification follow by one or more amplifications. This is also the last step. The signal is now ready to be digitized.[7]

In order to digitize analogue signals analogue-to-digital converters must be used. This converter measures the analogue signal so many times per second. This is expressed in samples per second (S/s). The most EEG signal measurements measure frequencies up to 100 Hertz (Hz). According to the Nyquist sampling criterion a signal should be sampled with a frequency at least twice the highest frequency of interest in order to determine the amplitude and frequency of the signal that was sampled. So as it turns out, a sample frequency of 200S/s is enough, but higher sample rates are also possible and several commonly used frequencies are 100, 250, 500, 1000 and 2000S/s. Furthermore, the quantization of EEG signals consists typically of 2^{16} levels, hence one sample is represented as a 16bit binary number.[7]

To have a good recording of EEG signals the electrodes should have an impedance less than 5 kΩ. In addition they should also be balanced to within 1 kΩ of each other.[7]

2.4 Positioning and Recording Modes

The International Federation of Societies for Electroencephalography and Clinical Neurophysiology have developed a standard for electrode placement. It is called the 10-20 system and is depicted in Figure 6. Left out on Figure 6 are the reference electrodes typically placed at the ears. In the past topographic distortion could be generated when the reference electrode was not relatively neutral. However in modern systems the placement of the reference electrode does not play a role anymore and other placements like F_{pz} can be used.[7]

The measurement of the EEG signals present at the different electrode positions can be measured in three ways:

- referential,
- differential,
- reference-free.

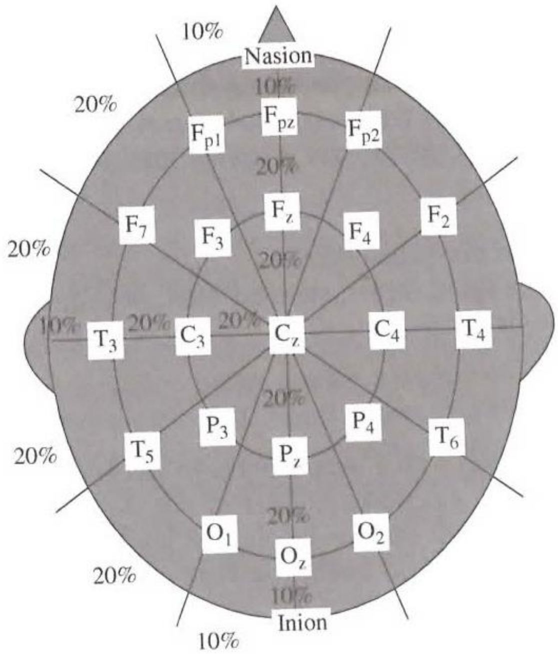

Figure 6: Electrode positioning consistent with the 10-20 system[7]

By measuring referential at a certain electrode one of the two inputs of the differential amplifier is the reference electrode. When measuring differential the inputs of the differential amplifier come from two different electrodes between which the difference signal is being measured. In the last method a signal is measured in reference to the average of the signal itself.[7]

According to [8] the differential approach has a benefit over the referential approach because it amplifies the minute difference of the signals between neighbouring electrodes which eases the difficult subsequent stages of amplification and data conversion.

3 The Sensor

This chapter fully describes the sensor. It commences by discussing the placement of the electrodes. Next, all the components are discussed. When all the components are discussed the part thereafter will try to explain how to connect the components.
The final part describes the program of the microcontroller. The format of the data that is being send is described separately. The different steps the program goes through is visually aided with a flowchart which can be found in section 3.4.4

3.1 Electrode Placement

As previously stated in section 2.1.2 an EEG signal originates from the cerebral cortex. Looking at predecessors (myZEO), sleep EEG is being measured at the frontal lobe or prefrontal cortex and frontal pole to be correct. Of course this is not blindfoldedly accepted but also [9] stated that experimental data indicates that the prefrontal cortex plays a role in mediating normal sleep physiology, dreaming and sleep-deprivation phenomena. Fortunately this is also the most comfortable place on the head and since myZEO already developed a very comfortable headband, as can be seen in Figure 7,it was used for the electrodes of this project. The electrodes consist of highly conductive and very soft silver-coated fabric.
The headband consists of three electrodes. The 10-20 system calls these electrode positions from left to right F_{p1}, F_{pz}, F_{p2}. These positions are encircled on Figure 8.
The electrodes of the headband are connected to the hardware through the use of plastic snap buttons (see Figure 9). The plastic ensures that only time varying signals reach the internal circuit of the sensor. By doing so the patient is electrically isolated from the circuit preventing him from high DC currents.
As previously mentioned, the electrode impedance should be below 5 kΩ. Therefore the headband uses highly conductive silver coated fabric. Since the resistance is below 1 kΩ they will of course also be balanced within 1 kΩ of each other.

Figure 7: Electrodes made of silver coated fabric

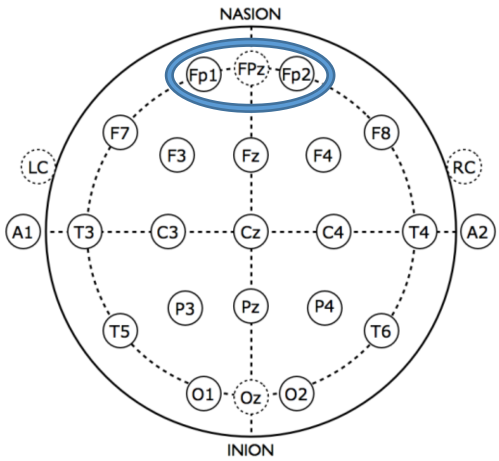

Figure 8: Positions used for this project[10]

Figure 9: The plastic snap buttons of the headband

Previous research ([8]) has proven that differential measurement makes the filtering, amplification and digital converting less critical. Consequently, the results would also be better. So it is only logical to choose the differential measurement for the development of the sensor. This however leaves one electrode remaining. In order to use all the electrodes, the one in the middle, F_{pz}, will be used as the reference node and the differential measurement will be between F_{p1}, F_{p2}. More about this in section 0 The Instrumentation Amplifier and section 0 Right Leg Drive.

3.2 Components[2]

3.2.1 Battery

The battery that is used to power the sensor is of the polymer lithium ion type and is depicted in Figure 10: The polymer lithium ion battery It was chosen as a function of its total capacity. As can be read in 6.1.1 the sensor's circuitry uses between 100-110mA. So in order to have a sensor that can last about 9 to 10 hours a capacity of 1000mAh is needed.
The battery was ordered from Sparkfun.[13]

The battery contains an internal circuit which protects the battery mainly from over-discharge and over- charge. The reader interested in a complete list of figures about the protection circuit is advised to read [11] and for a full list of figures about the cell itself [12]. Finally Table 2 sums up the most important specifications of the battery for this project.

Table 2: Battery specifications[13]

Output voltage	3.7V
Capactity	1000 mAh
Connector	2-pin JST-PH (2 mm spacing between pins)
Discharge current	1A (2A by cell but only 1A by connector)
HeightxWidthxLength	5.9 mmx33.5mmx50.8mm
Self-discharge	<8% per month
Temperature operating range	-25 to 60 °C

Figure 10: The polymer lithium ion battery[13]

[2] For a summary of all the components used please go to attachment A.

3.2.2 Battery charger/booster

The all-in-one charger booster displayed in Figure 11 is especially made for Lithium Polymer (LiPo) cells is used to make sure that the charging and discharging happens safely. The board is called the powercell board and is manufactured by Sparkfun.[14] It contains a charger IC(Integrated Circuit) MCP73831 and a booster IC TPS61200. Hence, it is a very small board with a size of 2.63mmX2.43mm.

Figure 11: All-in-one battery charger booster[14]

The powercell boosts the 3.7V of the LiPo towards 5V. This particular feature is handled by the TPS61200. In addition the powercell grants the possibility of shutting down the booster by breaking out the enable pin. By pulling it low the booster will be disabled and no power will be dissipated. In other words it can be used to shut down the sensor. This is very useful for when the sensor is not used.[14]

TPS61200 contains an under voltage lock out (UVLO) protection pin. The board has it set to be activated at 2.6V which is consistent with the battery specifications. The UVLO works as follows:
When the battery would discharge to far it would produce a voltage of 2.6V at the UVLO pin. If the battery would not be cut off from discharging it would get past its cut-off. At that point the battery is damaged and will not charge anymore. Therefore the UVLO pin will cut off the battery, making sure that it would not charge anymore. The UVLO protection comes in very handy when a person would not place the sensor back in its charging station[3] after a night sleep.[14]

The battery is charged either through USB or through an external power source connected to the +5V and ground (GND). The charging is controlled with the MCP73831 chip. Although the powercell contains a micro-USB it is not used. Of course there is a very persuasive reason for that. The fact is that the TPS61200 is not automatically shut off which means that the battery will keep discharging even though the sensor is not used. So instead of using the USB

---

[3] The charging station resides on the myRIO. Please see 4 Real-time NI Model myRIO for more information.

port to charge while pulling enable low it is much more user friendly to make a charging station which charges the battery through +5V and GND and pulls enable low eliminating the need of the USB charger.[14]

3.2.3 Negative power: Charge pump +5V to -5V

Some of the components used in the sections that follow need a negative voltage. This negative voltage is produced with the charge pump TC7660 which has a typical connection as shown in Figure 12.

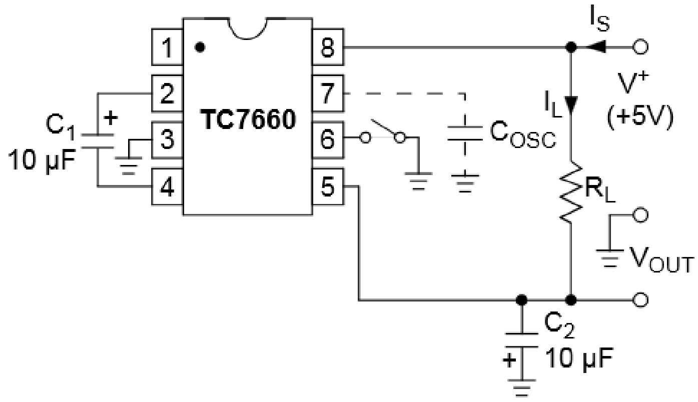

Figure 12: Connection of TC7660, a charge pump DC-to-DC converter[15]

3.2.4 Conditioning the Analogue Signal

Pre-filtering High Frequency Signals

The first stage of the analogue circuit does not amplify the signal but merely filters it from high frequency signals. One could speculate if the high frequency filter is an overkill since the leads are very short and in essence only present on a printed circuit board (PCB), thus not able to receive high frequency signals. Although that thought would not be wrong, it does not include the fact that humans are still very large antennas.
Sweat makes the skin conductive albeit being it very little. As a consequence the human body is susceptive to high frequency signals.

That leaves us to why high frequency signals need to be filtered separately especially since the next step in the analogue circuit is an instrumentation amplifier, a kind of differential amplifier.(see 0). They possess the ability to remove signals that are present on both electrodes such as high frequency signals.

James Bryant (ADI Europe) with Herman Gelbach (The Boeing Company) (www.analog.com/library/analogDialogue/Anniversary/14.html) describe the reason as follows:

> "An important factor is that, in instrumentation amplifiers, common-mode rejection decreases with increasing frequency, starting to roll off at quite low frequencies-and distortion increases with frequency. Thus, not only are high frequency common-mode signals not rejected; they are distorted, producing offsets".[16]

Furthermore James Bryant (ADI Europe) et al. (2014) explain that if the signal bandwidth is only a few Hz wide a simple RC(Resistor-Capacitor) low pass filter will remove the high frequency signals. Thank goodness EEG signals only have a small signal bandwidth. Figure 13 shows the implementation of the low pass filter. It consists out of three RC circuits. Two common-mode RC filters R12 * C2 and R13 * C3 which make sure the unwanted signals are filtered, and normal-mode RC filter (R12 + R13) * C1 which removes the unwanted high frequency signals from only one of the inputs.[16]

OpenEEG (http://openeeg.sourceforge.net/) is a website with open source information about an EEG system. They have developed a high frequency RC low pass filter as demonstrated by James Bryant (ADI Europe) with Herman Gelbach (The Boeing Company) (2014). The values of the resistors and capacitors obviously need to be the same since the bandwidth is the same, hence these values were used in the development of the sensor.[17]

Another feature of the protection circuit is the removal of high normal-mode voltages which are produced due to possible charge building on the sensors, they could also be produced by electrostatic discharges (ESD). In case these charges would not be dealt with a DC charge would amplified. Worst case scenario, the amplifier saturates leaving no room for EEG signals. Moreover, not only do these charges mess with the EEG measurement they could harm the circuit and more importantly the patient[4]. This feature is acquired through the use of transistors. However they could just as well be diodes. They operate as follows: if the voltage becomes higher than 0.7V they start to conduct. Consequently, the charge will flow away. The resulting circuit is depicted in Figure 13.

[4] That is in case someone would use electrodes which allow DC to pass.

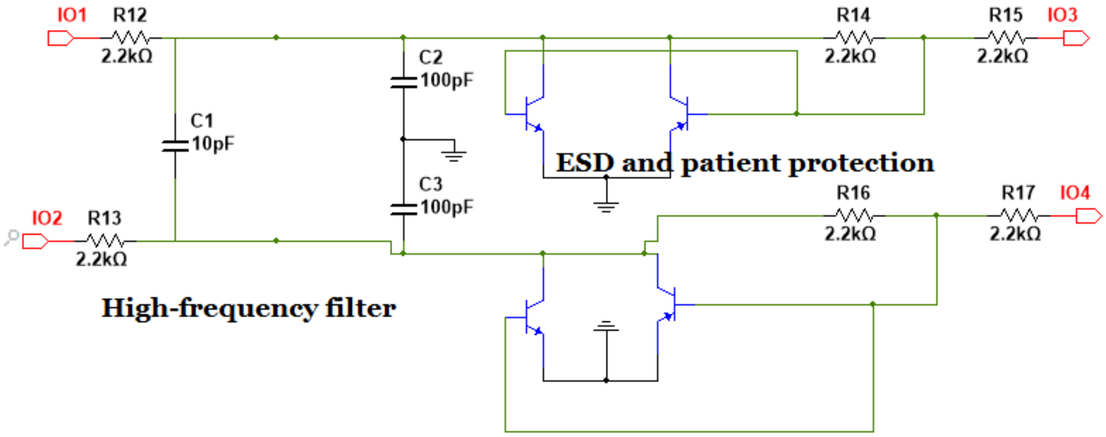

Figure 13: High frequency filter and ESD/patient protection circuit

The Instrumentation Amplifier

The first step in the amplification process is the differential amplification of the two electrodes. The component used is the INA 114PA from Texas Instruments, it is typically used for medical instrumentation. The schematic representation can be seen in Figure 14.

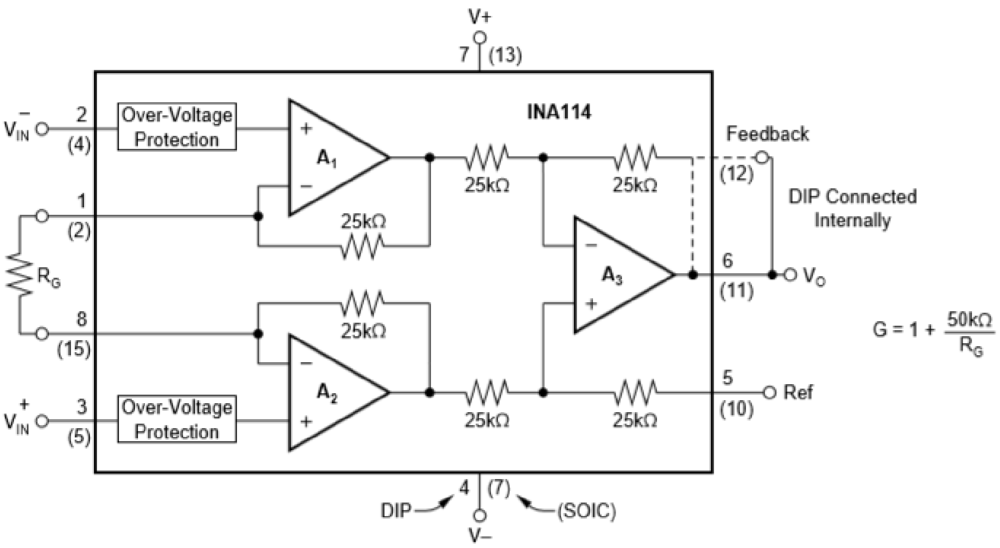

Figure 14: Schematic representation of INA114, right: the gain formula[18]

Table 3 lists the features which are of utmost importance for the measurement of EEG signals.

Table 3: INA114 specifications[18]

Offset voltage	50 µV
Drift	0.25µV/°C
CMR (common-mode rejection)	115dB
Minimum supply voltage	+/- 2.25V
Gain (G) (R_G is situated on the left of Figure 14)	$G = 1 + \frac{50k\Omega}{R_G}$

These features are important because the EEG signals are very small. So knowing that they range from 2 to 200 μV the offset voltage and drift should be as low as possible and the CMR (common-mode rejection) as high as possible. Furthermore, the power usage should be as small as possible, thus a low input voltage is preferable.

The instrumentation amplifier (INA) is powered with +/- 5V, but the gain however is only set to 4[5]. In fact the INA is used only to amplify somewhat but mainly to remove the common-mode. Moreover this common-mode signal is used as the right leg drive (RLD). More on this in the next section.

Refactoring the gain formula to R_G and substituting 4 for G the resistance between pin 1 and 8 should be 12.5kΩ. However, the common-mode needs to be measured at the point balanced between the two signal inputs. Hence, the resistance has to be split in two separate resistors as shown in Figure 15. Taking in to account which resistance values are available, the most suitable resistance value is 6.8kΩ. With this value the gain becomes 3.67.

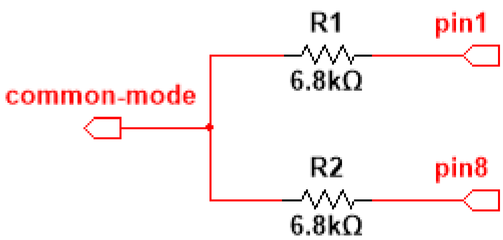

Figure 15: The common-mode output

[5] This is another preventive action against high offset voltages present at the input of the INA.

Right Leg Drive

The driven right leg is a method where the ground is driven to a known potential. By doing so the unwanted common-mode signals will be accounted for and removed from the signal[6]. Hence the CMR will increase even more.

The RLD is implemented as the "closed-loop RLD system" [19].It will drive the ground with the inverted common mode signal in order to cancel out the common-mode that was detected in the first place. Furthermore the RLD and INA are separated with a voltage follower, this creates a very small load for the INA, otherwise the resistance R_G "seen" by the INA would be different which would result in a different amplification . This circuit is depicted in Figure 16.

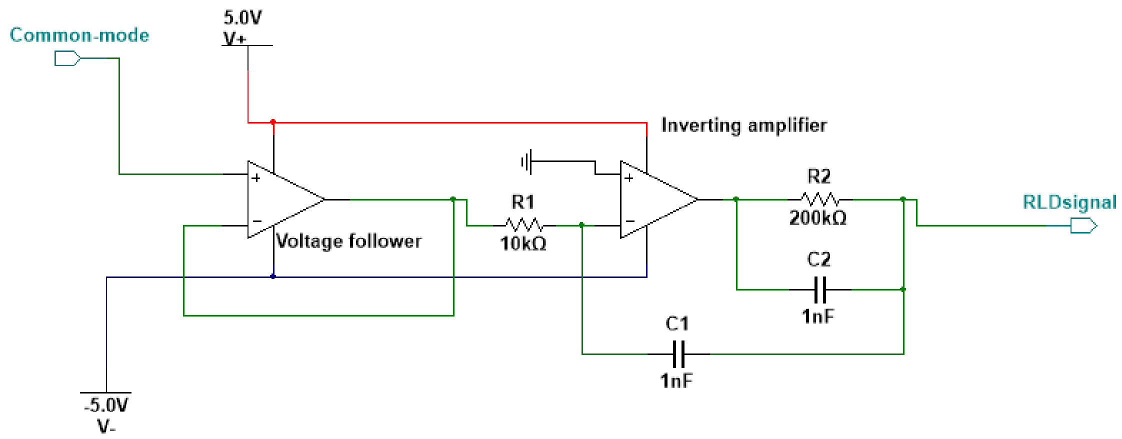

Figure 16: The Right leg driver circuit

The actual implementation is as implemented on openEEG (http://openeeg.sourceforge.net/) who also used the "closed-loop RLD system".[17] The RLD is implemented with the TLC277CP chip. It contains two operational amplifiers (OPAMP) in one IC. Indeed, the less space it takes the better.[20]

Subsequent Amplification and Filtering

Two steps remain before the ADC comes in to play. These two steps are exactly the same. A step consists of a high-pass filter[7] which removes disturbing DC offsets followed by a Sallen-Key implementation of a second order low pass filter which is illustrated in Figure 17. In order to save space- and keep the sensor's size as small as possible- the Sallen-Key also amplifies the signal 100 times. The values of the impedances have been calculated with a

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[6] The reader who is interested in the signals coupled on to the human body is advised to read [19].

[7] It is set with a RC high-pass filter circuit with a cut-off frequency of 0.16Hz. R=1MΩ and C=1μF.

simplified formula.[21] It sets the resistor values as a ratio while the capacitors are set with the same value.

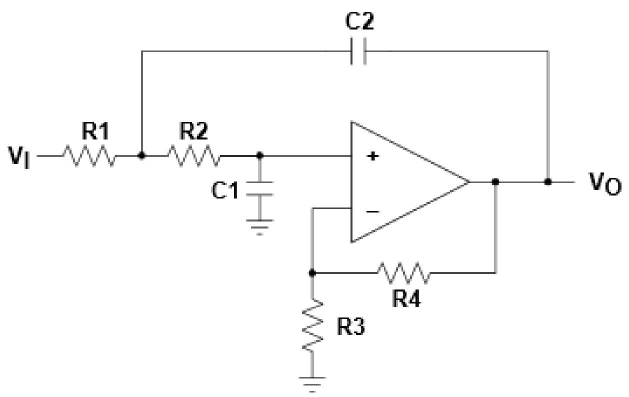

Figure 17: Second order Sallen-Key Low pass filter[21]

The ideal low pass Sallen-Key transfer functions are given in Equation 1 with K the amplification, fc the cut-off frequency and Q the quality factor of the filter.

With the proposed simplifications given in Equation 2 these equations become as given in Equation 3.

$$fc = \frac{1}{2\pi * \sqrt{R1 * R2 * C1 * C2}}, \qquad Q = \frac{\sqrt{R1 * R2 * C1 * C2}}{R1 * C1 + R2 * C1 + R1 * C2 * (1 - K)}$$

Equation 1: The ideal low pass Sallen-Key transfer functions[21]

$$R1 = m * R, \qquad R2 = R, \qquad C1 = C2 = C$$

Equation 2: The established simplifications[21]

$$fc = \frac{1}{2\pi * R * C * \sqrt{m}}, \qquad Q = \frac{\sqrt{m}}{1 + 2 * m - m * K}$$

Equation 3: The simplified ideal low pass Sallen-Key transfer functions[21]

Care was taken when choosing the values of the capacity and resistances. Indeed, there is interaction between fc and Q due to the common parameter m. [21] suggests starting the design with choosing m and K to set the amplification and quality factor. From here the capacity should be chosen and finally the resistance R should be calculated with the required cut-off frequency.

The implementation requires a gain of 100 and a cut-off frequency at 80Hz. Q just needs to be positive.

The complete calculation is quite lengthy, therefore it has been moved to attachment B.

Computing the values of the impedances and taking the availability into account, the results are as presented in Table 4.

Table 4: Impedance values for low pass Sallen-Key

Physical quantity	Value
R1	100Ω
R2	82kΩ
R3	150Ω
R4	15kΩ
C1	0.68µF
C2	0.68µF
K	100
fc	81.73Hz
Q	0.034

Notice that Q is relatively small. Therefore care should be taken with the deviations of the impedances. If these become too large, it could get the system to oscillate.

The two OPAMPs needed are also implemented with the TLC277CP chip.[20]

The signal is now ready to be converted into a digital signal and send to the central processing unit. The Bluetooth modem will be enlightened first.

3.2.5 The Bluetooth Modem

Figure 18 presents the Bluetooth modem which is used in the composition of the sensor, but also on the myRIO. It is called the BlueSmiRF Silver and was ordered from Sparkfun.[6] Its most important settings are represented in Table 5.

Figure 18: Bluetooth Modem BlueSMiRF Silver[6]

When a connection is made to another Bluetooth device this Bluetooth modem will send a string over UART saying that is connected successfully. That way the microcontroller as well as the myRIO is able to tell that a connection was established by reading this string.
Same thing for when the device disconnects. Then it will send a string indicating that it has lost its connection.

Table 5: Bluetooth modem settings

Feature	Setting
Bluetooth address	000666715B10
Bluetooth name	MyRIOBt
Baud rate	115200
Data rate	8 bits
Parity	None
Stop bits	1
Status string	%ESC

In order to send the measured EEG signals the microcontroller has to write the data in the form of strings to the Bluetooth modem and of course the myRIO has to read data strings from the Bluetooth in order to receive the EEG signals.

No level shifter is needed for the pins as they are all 3-6V tolerant.[6]

3.2.6 The Microcontroller

The microcontroller used in the sensor is called the Teensy 3.1 and was ordered from Sparkfun.[22][5] There is not a single downside to using this microcontroller. It is small, has a lot of RAM (Random Access Memory) and has a built in UART port. It is programmable in C or in the Arduino IDE. The latter was used in this project. Figure 19 displays the teensy controller, Figure 20 the pin assignments where the pins that are used are marked with a red cross. Their function is underlined, but here follows a nice little overview:

- UART port on pin Rx1 and Tx1,
- ADC of EEG signal on pin A9,
- temperature value on pin D12,
- the power supply of the Teensy on Vin,
- 3.3V power supply from the Teensy to temperature sensor,
- and the ground of the teensy on both GND pins.

Figure 19: Microcontroller Teensy 3.1[5]

The Teensy has three important functions:

- making the Bluetooth connection,
- measuring the filtered EEG signals,
- measuring the temperature,
- sending the information.

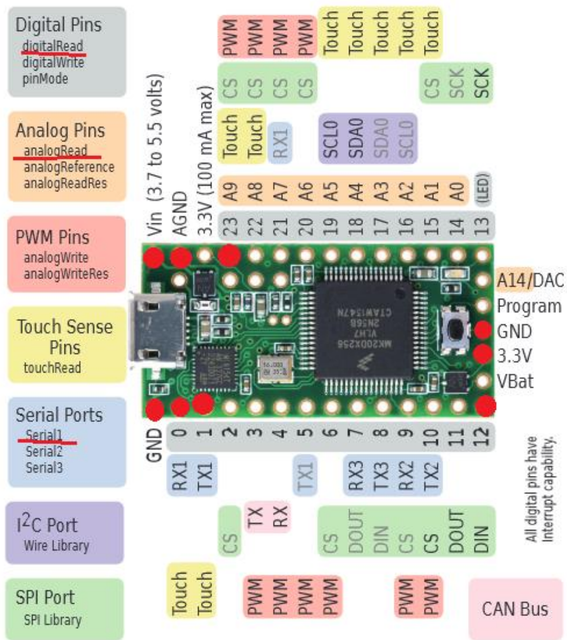

Figure 20: Teensy 3.1 pin assignments[22]

The Teensy can only convert signals between 0 and 5 volt. However the conditioned signal ranges from -5 to 5 V. Therefore the signal needs to be converted to the 0 to 5 voltage range. This can be achieved by using the circuit that is shown in Figure 21.

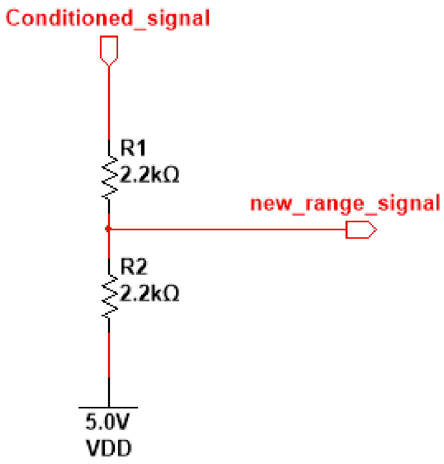

Figure 21: Range transformer

Assume three cases. The conditioned signal is -5V, 0V or 5V.

In the first case when the conditioned signal is -5V there will be a potential of 10V over the two resistors in series. The 10V will be equally divided between the resistors, thus they both have 5 volt across it. That means that the node in the middle has a voltage of 0V. Indeed, VDD-5V.

This calculation also counts for the second and third case. The node will have a voltage of 2.5 and 5 respectively.

In conclusion the voltage measurable at the output will range from 0 to 5V.

Although the literature suggested that sampling typically happens with 16 bit ADCs the Teensy will convert the signal to 13 bit values although it has the capability of 16 bit conversion. However, due to noise present in the signal it is limited to 13 bits. [23]
The sample frequency is set to 250 S/s.

3.2.7 Extra: Temperature Measurement

The sensor contains a DS18B20 temperature sensor.[24] It does not take up a lot of space as can be seen in Figure 22. The connection circuit is depicted in Figure 23. It is implemented with a 10 bit temperature measurement, although the communication protocol is I2C(Inter-Integrated Circuit)[8] the component is controlled with a prewritten library which provides easy methods to communicate with the temperature sensor. That library is called <DallasTemperature.h>.[25]

This component was ordered from Sparkfun.[26]

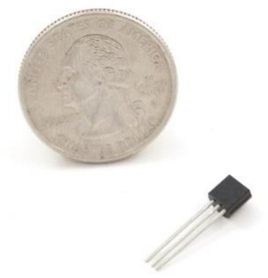

Figure 22: DS18B20 temperature sensor[26]

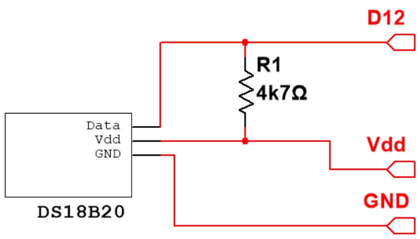

Figure 23: DS18B20 connection

[8] Developed by NXP Semiconductors N.V.

3.3 Hooking up

The sensor consists of two single sided printed circuit boards (PCBs) which have the Teensy and LiPo powercell sandwiched in between. On the first PCB resides:

- the BlueSMiRF Bluetooth module,
- the Teensy,
- the charge-pump,
- and the powercell.

This PCB than connects at several points with the other PCB which contains the rest of the hardware, which is the analogue circuit. This PCB can be seen in attachment C.

There is not much to tell about drawing traces on a PCB nor does it increase the clarity, on the contrary, it is indistinct. Nevertheless the design of the first PCB is shown in Figure 24 in order to explain several specific features when connecting to the charging station.

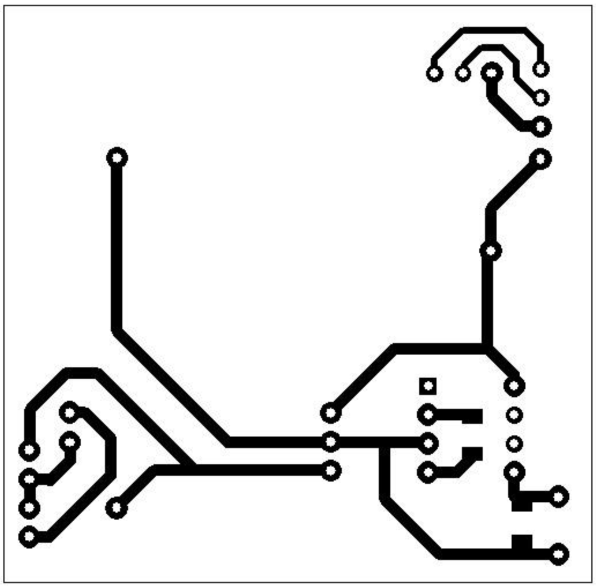

Figure 24: The first PCB

On the top right vertically below each other are the pinholes for the Bluetooth sensor.
To the left of this are three pinholes. These are from the right to the left: the GND pin of the microcontroller, the RX1 pin and the TX1 pin.
Completely to the left is the GND pin which comes from the power source.
To the right in the middle, in the middle of a track is the voltage in of the Teensy, it comes from the power source.
The eight organised pinholes connect to the charge-pump and the little rectangles are the solder pads for the two 10µF capacitors. From the two bottom pins to the lower right the highest one delivers the -5V the lowest one is the GND.
The three pads vertically below each other connect to the powercell. From high to low these pads connect to: the Vcc pin of the powercell, the GND pin and the enable pin.

Finally now comes the interesting and important feature when connecting to the charging station.[9]

On the lower left there are 4 pads. The lowest one is connected to an external 5V source and the copper track leads it towards the 5V charging source of the powercell.
One pad higher is the external GND. It goes to the next pad above it. This pad goes to the GND charging pin of the powercell.
On the same horizontal level as the external GND pinhole another pad can be seen to the right. This one also connects to the external GND. It connects to the enable of the powercell making sure that it does not dissipate power.
However this would only work if the myRIO was powered, otherwise the battery would still discharge. To prevent this the upper pad on the lower left is connected to the one below it when the sensor is placed onto the charging station. Consequently the enable would be connected to the GND charging pin. And since this ground is always active the powercell would pull its own enable pin low thus shutting it off.

[9] People who skipped 3.2.2Battery charger/booster are advised to read it in order to understand the specific feature explained.

3.4 Microcontroller Software

This section gives an overview of the software that is written for the microcontroller. It will be explained through the use of a flowchart. However, the formation of the string that contains the data and which will be send will be explained individually. Readers interested in the complete code can find it in attachment D.

3.4.1 The Flow Step-by-step

This section will explain every step in the program. For a visual aid the reader can go to section 3.4.4 where Figure 25 the flowchart of the microcontroller software is situated.

Initialise

The program applies some general settings which do not change during the program. Therefore they only need to be set once.

Connect

Once the general settings have been applied the program waits for an incoming connection. The acknowledge the program is waiting for is the string ESC%CONNECT,MACADDRESS,0.

Once this acknowledge has been received the MAC (Media Access Control) address will be stored and the program moves on.

Wait for Setup

As soon as a connection has been established the program is waiting to receive the setup from the myRIO. If the microcontroller has not received the setup before the watchdog timer[10] goes of the sensor disconnects itself and goes back to waiting for a new incoming connection.

If the setup was received the settings are applied and the program continues to the main loop.

Currently the physician can only choose whether to also measure the temperature or not.

[10] Watchdog timer: A timer that induces a specific action if it flows over. It can be reset before it flows over in order to prevent the action from being induced.

Main Loop

In the main loop 4 things are being monitored:

- a watchdog timer that resets when signals are received;
- a timer that starts an EEG measurement when it flows over;
- a timer that start a temperature measurement when it flows over;
- if the char 'X' is received.

This can be seen on the flowchart in Figure 25 where 4 arrows are leaving the main loop state.

In case the X is received the sensor gets reset and returns to the connect state. It is than able to connect with another device.

Reconnect

The microcontroller goes into this state when the watchdog timer in the main loop went off. It is not the microcontroller itself that is trying to reconnect. The microcontroller merely waits for an incoming connection of the myRIO which is of course also trying to reconnect.

The MAC address incorporated into the acknowledge is compared to the one stored. If it is not the same than the sensor will try to disconnect from this device and it will not send any signals to it.

EEG

This state measures and sends the EEG signals. A measurement is taken every 4 millisecond, hence the frequency is 250Hz. For more information about how the data is transmitted please go to 3.4.2.

Temperature

This state measures and sends the EEG signals. A measurement is taken every minute. For more information about how the data is transmitted please go to 3.4.2.

3.4.2 Building the Transmitted String.

The measurements made by the ADC are formatted in to a string and send to the Bluetooth module that sends them to the myRIO.

In the first step measured data is split into two parts: the two highest digits and the two lowest digits. Immediately after that the current millisecond timer value is stored. It consists out of 8 digits and is split into 4 groups.

In the following step all these groups consisting out of two digits are converted to bytes.

Right before the data is send it gets a header and a trailer. The header is a byte that tells whether the data is a temperature measurement or EEG measurement. The trailer is the delimiter which tells myRIO that the string is complete.

These bytes are send as ASCII (American Standard Code for Information Interchange) characters. As a consequence the data could contain the same ASCII character which defines the delimiter. Therefore the most significant bit of the data bytes are set to one; after all that bit is never used since two digits can only go up to 99 and that means that 7 bits are enough rendering the last bit unused. By setting the most significant bit high the delimiter value of 000 0013 is never reached, thus solving our problem.

The following section gives an overview of the bytes that are send.

Format of the String

THLM4M3M2M1\n

T: Specifies whether this string contains temperature data or EEG data.
  B01101101 for EEG data
  B01111010 for temperature data

H: The highest two digits of the measured data.
L: The lowest two digits of the measured data.
M4: The highest two digits of the millisecond timer.
M3: The third and fourth highest digits of the millisecond timer.
M2: The fifth and sixth highest digits of the millisecond timer.
M1: The lowest two digits of the millisecond timer.
\n: This is the delimiter which specifies that the data has been fully received.

Example of a Formatted String

An example will clarify things up. Keep in mind that this example shows the data before it is decoded into bytes.

m320310254616

m= B01101101 thus EEG data
H=32
L=03
M4=10
M3=25
M2=46
M1=16

3.4.3 Libraries

Three different libraries were used. They will shortly be described.

<OneWire.h>:

> The library that facilitates the communication with one wire protocol sensors. This library is only used by the <DallasTemperature.h> library.[27]

<DallasTemperature.h>

> The library that communicates specifically with the DS18B20.[25]

<TimedAction.h>

> Library that triggers an event every X milliseconds. In this case the measuring of the temperature or EEG.[28]

3.4.4 Program: flowchart

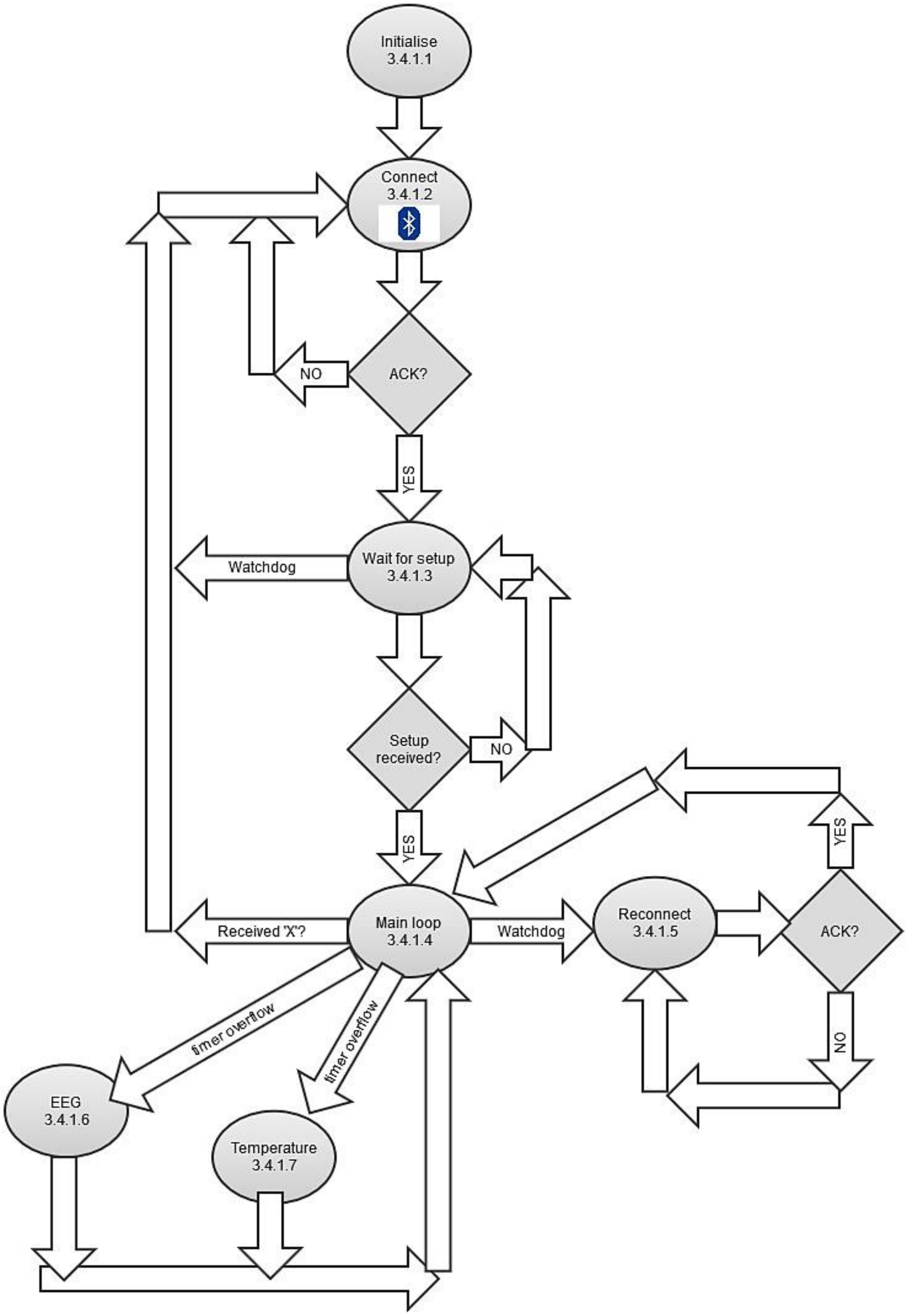

Figure 25: Flowchart of the microcontroller program

4 Real-time NI Model myRIO

This chapter discusses the properties of the myRIO that are most important to this project.
The first part explains how components can interact with the myRIO.
The second part discusses the storing of data onto a USB flash drive.
The third part discusses the use and working principle of network shared variables.
The last part discusses some minor inconveniences of the myRIO.

4.1 MXP Protoboard

MXP stands for myRIO expansion port. It is this port which contains the UART port to interface with the Bluetooth modem, a digital IO pin to detect the charging of the sensor and 5V output to charge the sensor, a PWM output to drive a buzzer and also an I2C port to read the date and time of the DS1307 I2C chip. National Instruments attached a protoboard to the MXP port so components could interface with it nicely.[29] Figure 26 only serves for illustrative purposes but it shows nicely the components connected with the MXP protoboard.

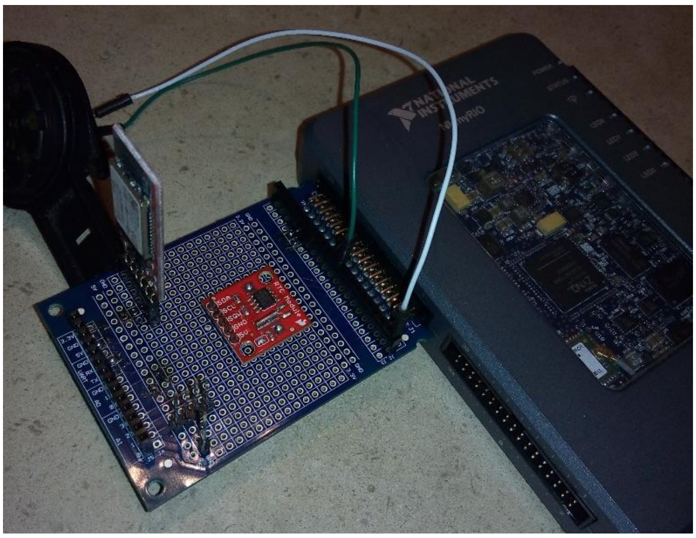

Figure 26: Components of MXP board

4.2 USB as a Storage Device

A USB port is incorporated into the myRIO. It allows for files to be stored and accessed programmatically.

4.3 Shared Variables

As far as the real-time interface for the physician is concerned, shared variables play the most important role. It makes data on the myRIO available through a network, has not to be wireless per se, but since it is possible via Wi-Fi it creates the real magic and makes a wireless interface available throughout the same network. Shared variables or network shared variables actually, are variables deployed from a shared variable engine (SVE) on a certain NI system, in this case the myRIO. Hosts can access the SVE and request specific variables. Once a new value has been written to a variable it gets updated at every subscriber.[3] This is depicted in Figure 27.

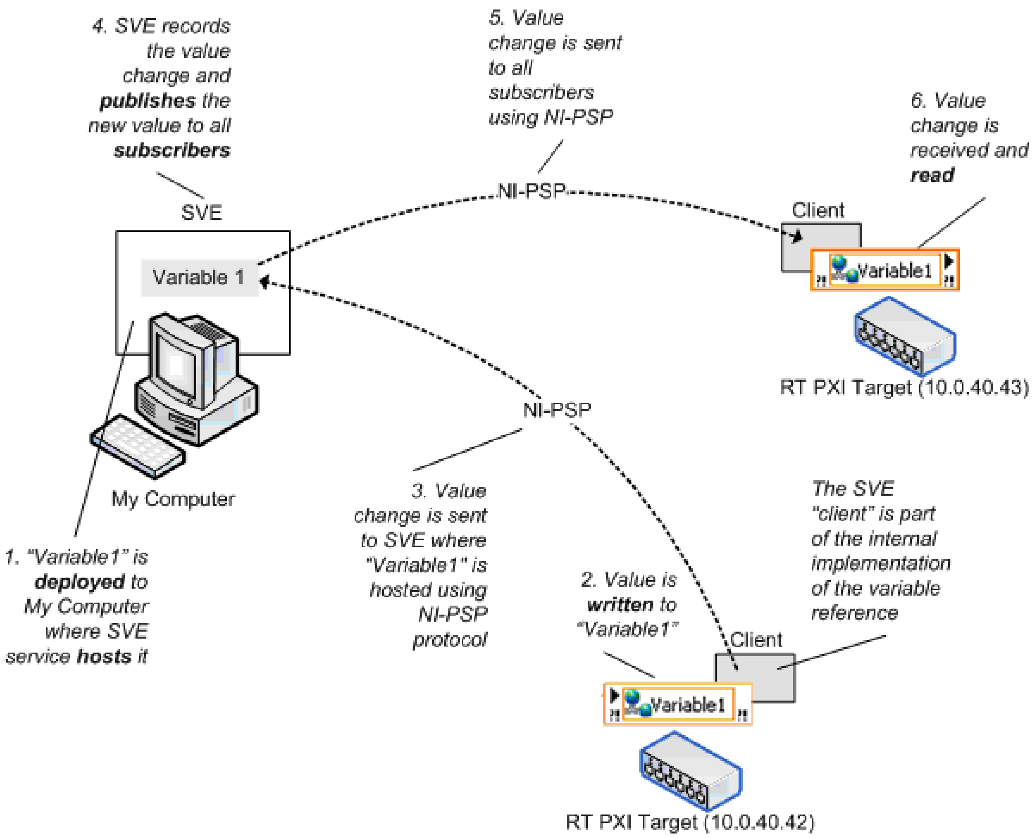

Figure 27: Update sequence[3]

Furthermore every host receives its own buffer. This is illustrated in Figure 28. For example: the current settings contain a setting for the rating range. It could be from 0 to 4 or from 0 to 10. This setting is send with a shared variable to the host computer and to the registration application which both receive the value and both can read it at a different time since they both have their own buffer.

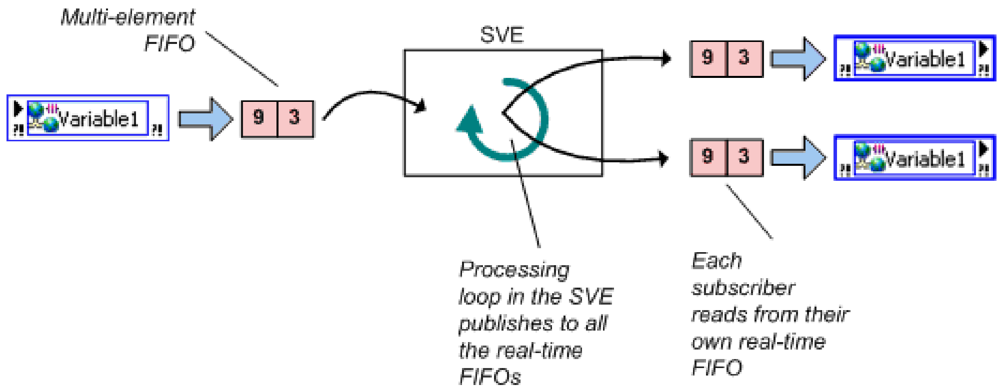

Figure 28: Separate host buffers[3]

4.4 Some Minor Issues

Although the myRIO has its own NI Linux Real-Time (ARM-based) Operating System , it only has a limited set of bash functions that can be used. For instance it is not possible to ping another device in the same network let alone just checking wetter you have internet connection[11]. This problem could not be resolved but precautions were taken like the button to stop the beeping if the internet connection would be lost.

Another minor inconvenience is not an inconvenience anymore as it has been solved with an external component namely the real-time clock module with the DS1307 I2C chip as seen in

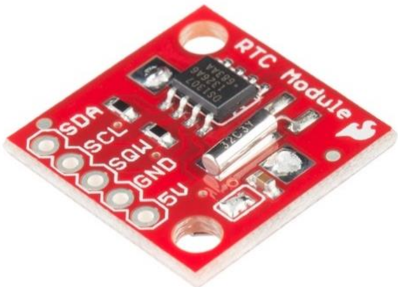

Figure 29. Indeed, the myRIO has no internal timekeeping, so whenever the myRIO disconnects from power time figuratively freezes for it and resumes when there is power again. By using the real-time clock module the date and time on the myRIO's OS will be adjusted with the date and time read from the DS1307 chip. The real-time clock module has a battery and can run up to 9 years. It was produced by and ordered from Sparkfun.[30]

Figure 29: The real-time clock module[30]

A small downside when using shared variables for the first time in a new network: the network shared variables need to be reconnected. But just once.

[11] At least when no host is connected to the myRIO's shared variables. Otherwise one could set a shared variable, connect it to another one in the host's program and read this one on the myRIO. If the test is positive you have a connection.

5 Processing with LabVIEW

This chapter describes the LabView software. There are two programs written in LabView: the graphical user interface and the program on the myRIO.
The graphical user interface software is described in the last part and is not all that large and is therefore described in only a few sections.
The program on the myRIO however contains several programming structures. Their purpose will be explained in full in the first section.
The main finite-state machine is visually aided with a flowchart that can be found in section 5.1.4.

5.1 myRIO Software

The software consists out of three parallel loops. That way multiple processes can run at the same time. The simultaneous running of these processes are necessary because the myRIO should be able to receive data from the sensor and the registration interface at the same time. If this would be implemented in the same while loop as the measurements, the measurements would stop once a rating is requested and only recommence once the rating has been given.

5.1.1 The Shared Variables Loop

This loop is a collection of the shared variables that are being used in the main finite-state machine. By placing them into a separate while loop they will be continuously updated.

5.1.2 The Registration Loop

This loop receives all incoming data from the rating registration. It is able to differentiate between intentional awakenings and spontaneous awakenings. Furthermore it is also able to detected whether the sleeper has given an out of range value. This could occur when the sleeper enters a 6 when the rating is only from 0 to 4.

It is also this loop that controls the beeper.

5.1.3 The Finite-state Machine

The finite-state machine can be interpreted as the heart of the whole system. In 5.1.4(Figure 30) the finite-state machine is presented. Notice that every state could go to the end state in the next state. This happens in case of an error, reboot or docking of the sensor, which basically means the end of the program.

In total there are 7 states, they will now be explained individually.

Set Date and Time

As previously mentioned the myRIO has no internal timekeeping so as soon as power is lost time freezes. Indeed, the system should set its internal clock when power is applied. The correct time is read from the RTC module[30]. The protocol used to read the date and time is I^2C.

Settings

In the settings state several steps occur. In the first step the presence of a USB device is checked and in the next the settings file present on the USB is read. If no settings file is found a standard file will be generated. From these settings the size of the generated data from one night is estimated plus they are send to the graphical user interface (GUI). If it is larger than the available space on the USB device the program will not start. If there is enough room left, the myRIO waits for the user to start the program. This can either be done by picking up the sensor, pushing the button on the myRIO or via the start button on the GUI. When the program is started the myRIO goes immediately to the next state connection.

Some additional features are:

- LED1(light emitting diode) on the myRIO indicates if there is no USB present;
- LED2 indicates if the available space on the USB device is insufficient;
- as long as the program is not started the settings can be changed;
- the hold button on the GUI can prevent the program from starting as long as the button is set to true;
- creation of the file that saves ratings that were given spontaneously.

Connection

When the myRIO is in the connecting state it depends on the connection mode setting what this state will perform.

If it is the choose mode the myRIO will first look for every available EEG sensor. The myRIO connects very briefly with them to see whether they send "EEGD" which is an acknowledge of being a EEG measuring device.
All devices that pass the test are send to the GUI so that the user is able to choose one of them. Then in the next iteration of the while loop in this state the myRIO will try to connect

to the chosen device. If the connection cannot be established the myRIO sends an error report and sends a new list of available devices.

In auto mode the myRIO performs the same process as in choosing mode, but instead of letting the user chose the myRIO will connect to the first device in the list. If this connection would fail then the program will try the next device and so on until all the devices have been tried. Then an error message will be sent.

In both choosing and automatic mode an error message will be sent if no compatible devices have been found in the first place.

Of course when the bedside mode is intended to be used than the connection mode should never be set to choose otherwise the program will get stuck from the moment the myRIO expects a device to be chosen.

Finally the last connection mode is the previous mode. In this mode the myRIO connects to the last device that was successfully connected. This MAC address is stored onto the flash-memory of the Bluetooth module.

When errors occur in the bedside mode the program just keeps retrying with the selected connection mode.

Setup

If a successful connection was established the program goes to the next state, the setup.

In this state a file that saves all measurements and a file that saves all ratings from intentional awakenings is created.

Parallel to that the setup for the EEG sensor is being send.

Measuring

This state consists out of a producer consumer loop. The producer loop receives the data and decodes it back to the original signal. Finally the decoded sample is enqueued. Furthermore the time between subsequent incoming data is monitored. Should this time exceed some kind of limit than the myRIO considers this connection broken and the myRIO goes into the reconnect state.

The consumer dequeues the samples and adds them to a FIFO register. The different spectrums that were requested are formed and saved into a csv file. In parallel to that the trigger points are also monitored and acted upon if the conditions are fulfilled. Acted upon means that the trigger point is saved. The subsequent rating from the sleeper is saved in the registration loop. Should however the sleeper not react in time then the consumer loop will add “No rating received!” to the last trigger point.

Reconnect

In case the connection is lost the program enters this state and tries to reconnect with the sensor. Upon successful reconnection the state machine returns to the measuring state.

End State

This is the state that ends the program. All shared variables and settings are reset and the connection with the sensor will be terminated. The next state is the settings state.

5.1.4 The Finite-state Machine: Flowchart

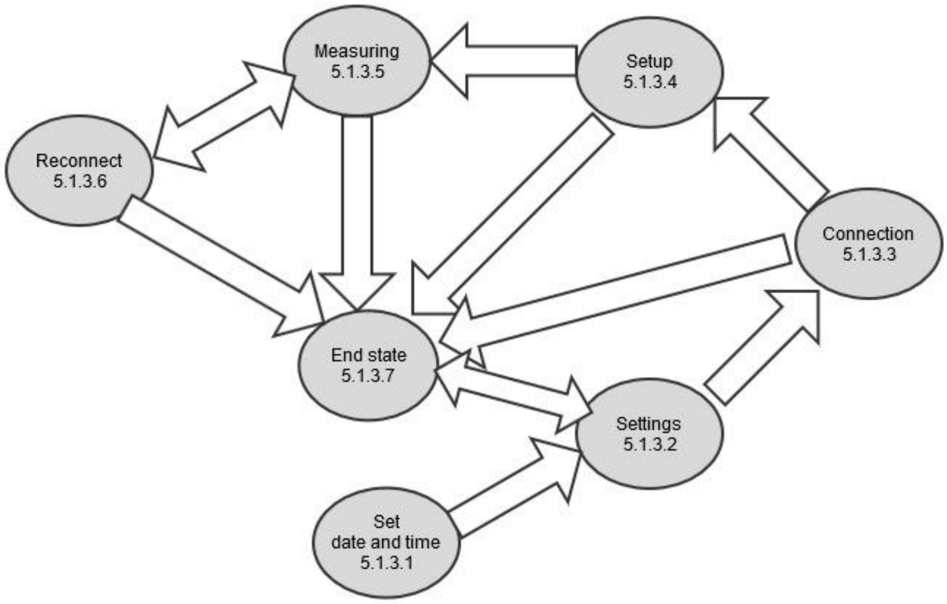

Figure 30: The finite-state machine flowchart

5.2 Interface Software

This software consists out of three parallel while loops. This is necessary in order to read the received data from the sensor continuously apart from what the other loop, which contains an event structure that reacts to events from the user, is doing. The remaining loop contains a collection of the shared variables that are used.

5.2.1 The Shared Variables Loop

This loop contains all the shared variables that need to be continuously updated. For instance, if a trigger point would be activated this should also be visible on the user interface and since an activated trigger point is important it should happen immediately.

5.2.2 The EEG Data Loop

The displaying of the EEG data should happen continuously. Imagine that this would not be the case. Then the EEG data would be read in one of the event cases. If then another event would occur apart from changing the tabs[12] the graphs would not be updated anymore although they are still visible.

5.2.3 The Event Structure Loop

This loop handles the interaction with the user. The events that are being monitored are:

- the activity on the tabs,
- pressing the connect button to connect to a specific sensor;
- buttons that go through the listed trigger points and spectrums;
- saving and updating settings,
- the deleting of trigger point and spectrum entries;
- and the selection of the spectrum to be shown.

[12] For more information about the interface go to 6.3.2The Physician's Real-time Interface.

6 Results

This chapter displays all the results starting with the behaviour of the sensor. The next section summarizes all the files that are being generated.
And finally the last section shows all the possibilities of the real-time interface as well as all the possibilities of the rating registration app.

6.1 Behaviour of the Sensor

6.1.1 Hardware

The sensor uses between 100 and 110mA when in use. That means that the sensor would last about 9 to 10 hours with the battery of 1000mAh that is used.

6.1.2 Software

The software on the sensor was tested with simulated signals.

Sometimes the connection failed and needed to be restarted but this should be handled by the myRIO.

The following test were performed and were positive:

- Rebooting the sensor,
- Loss of connection,
- Sending data,
- Trying to connect the sensor with another device during loss of connection (this is not allowed);
- Sending the valid connection char.

The following is an example of the received data that was transmitted.

```
m\C6\D0\80\97\9F\CF\n
m\C8\A8\80\97\9F\D3\n
m\C8\D8\80\97\9F\D7\n
m\C8\84\80\97\9F\DB\n
m\C9\B5\80\97\9F\DF\n
m\C9\CB\80\97\9F\E3\n
m\C8\9D\80\97\A0\83\n
m\C6\B2\80\97\A0\87\n
z\80\80\80\97\9E\B6\n
m\C7\9F\80\97\A0\8B\n
m\C7\E2\80\97\A0\8F\n
m\C7\CD\80\97\A0\93\n
m\C8\C4\80\97\A0\97\n
m\C7\DC\80\97\A0\9B\n
m\C6\94\80\97\A0\9F\n
m\C7\BD\80\97\A0\A3\n
```

It always consists of 8 chars what was expected. The m is the type byte of EEG data and the z for temperature data.

6.2 Data files

The name of these files represent the day and time they were made.

6.2.1 Spontaneous Ratings

This file saves:

- the time,
- the rating,
- and the comment (if one was given).

6.2.2 Triggered Ratings

This file saves:

- the conditions that were set,
- the time,
- the rating,
- and the comment (if one was given).

6.2.3 Setting Files

This file saves all the settings for a specific day. For a list of all possible settings, please got to 6.4.2.

6.2.4 Measurement Files

This file saves data in a csv file the following way:

- the start of the measurement,
- the value of the millisecond timer is saved in one column,
- one column always contains the raw data,
- every frequency spectrum and single frequency is saved in a separate column.

6.3 Interfaces

6.3.1 Rating Registration Interface

Figure 31 shows the interface made for the registration of a rating upon an intentional awakening. The interface is made with the NI Data Dashboard application for LabView.[31]

Patients receive a notification and they are able to give a rating between 0 and 4 or between 0 and 10. They can also give an optional comment. Once they have entered their data they hit send to save the rating. In addition they are never able to enter a number higher than 10. They can however enter a number higher than 4 when a rating between 0 and 4 is required. When that happens the beeper present on the protoboard will go off indicating that the rating was in the wrong range.

A person can enter a rating whenever he needs to. This includes before, during and after the night.

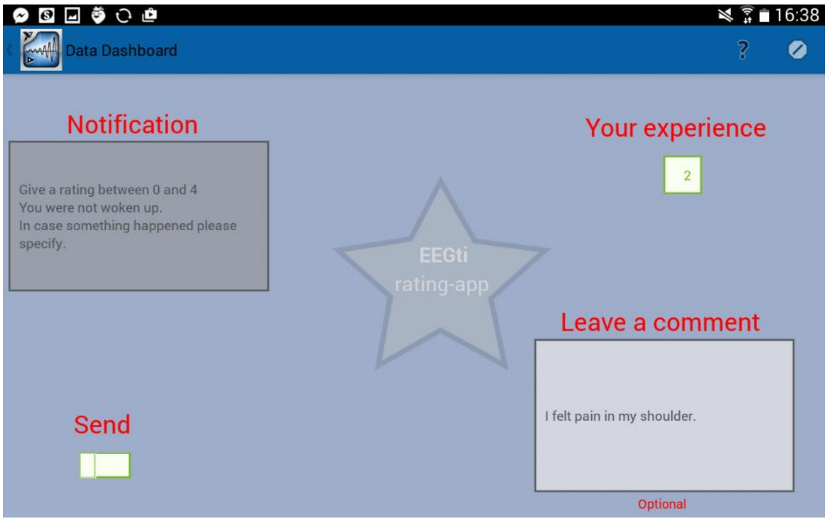

Figure 31: Registration interface

Should it happen that the myRIO is temporarily disconnected from the network, the application will show exclamation marks. Figure 32 shows this. This informs the person that the application is not able to register his rating at this point. To end the buzzing which went off in case of an intentional awakening, the person has to push the button present on the myRIO. Not a problem because the myRIO always has to be within Bluetooth range.

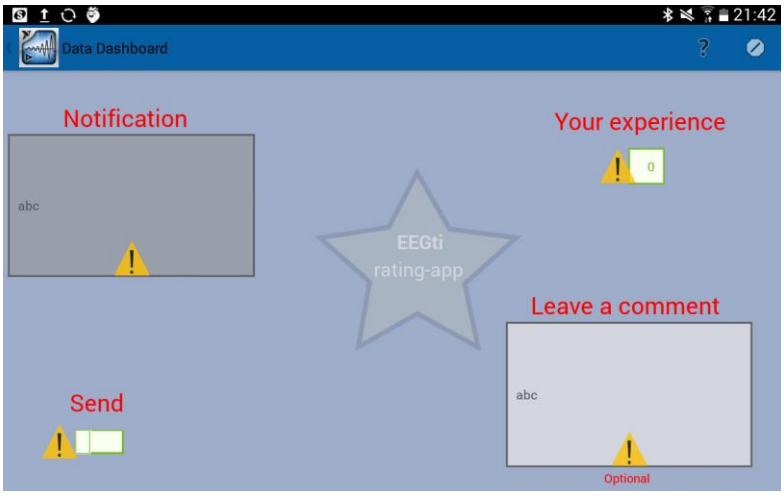

Figure 32: Registration interface unable to connect to myRIO

6.3.2 The Physician’s Real-time Interface

This section will explain all the features the real-time interface possesses. Figure 33 shows the interface upon start.

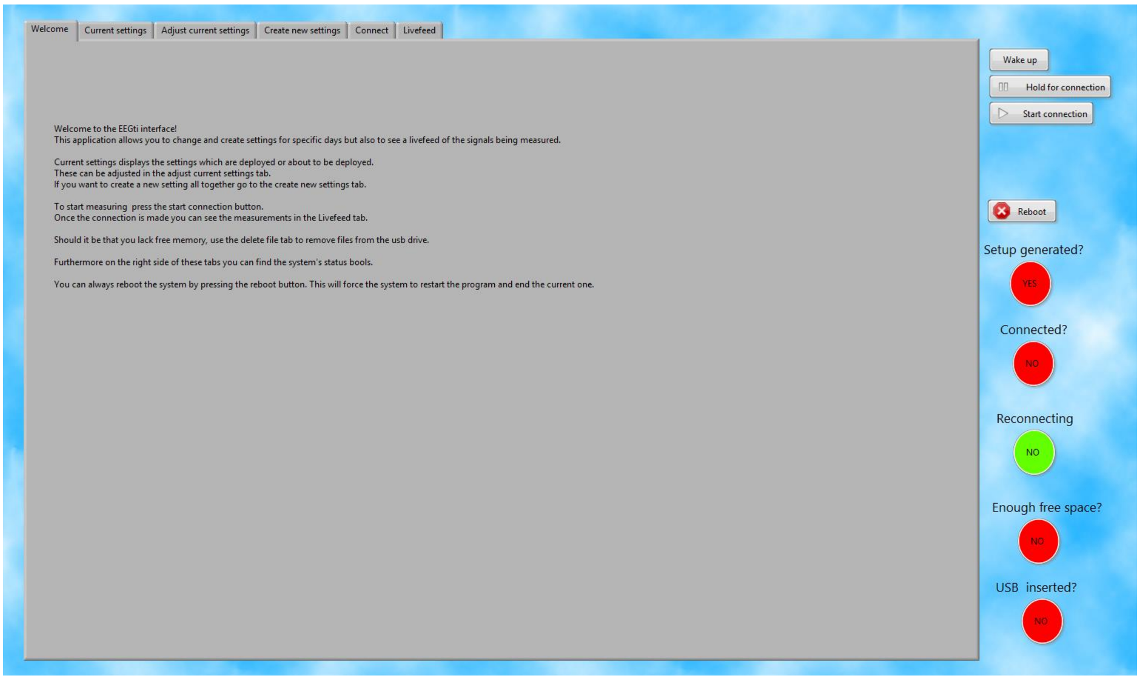

Figure 33: The welcome tab

Also, notice the buttons and indicators to the left. The indicators describe which data they represent. From top to bottom the buttons perform:

- the activation of a wake up signal,
- preventing the system from connecting to the sensor should it have been removed from its charging station;
- the start of a connection if there would be a different sensor;
- a reboot of the system in order to restart the system from the beginning.

The interface consists of different tabs. Each tab represents a certain function or visualization. These are laid out in the following sections

6.3.3 Deployed Settings

On the next tab, current settings, displayed in Figure 34, the settings that were loaded for the day in question are displayed. The user can go through all the trigger points and frequency regions, but they cannot be changed

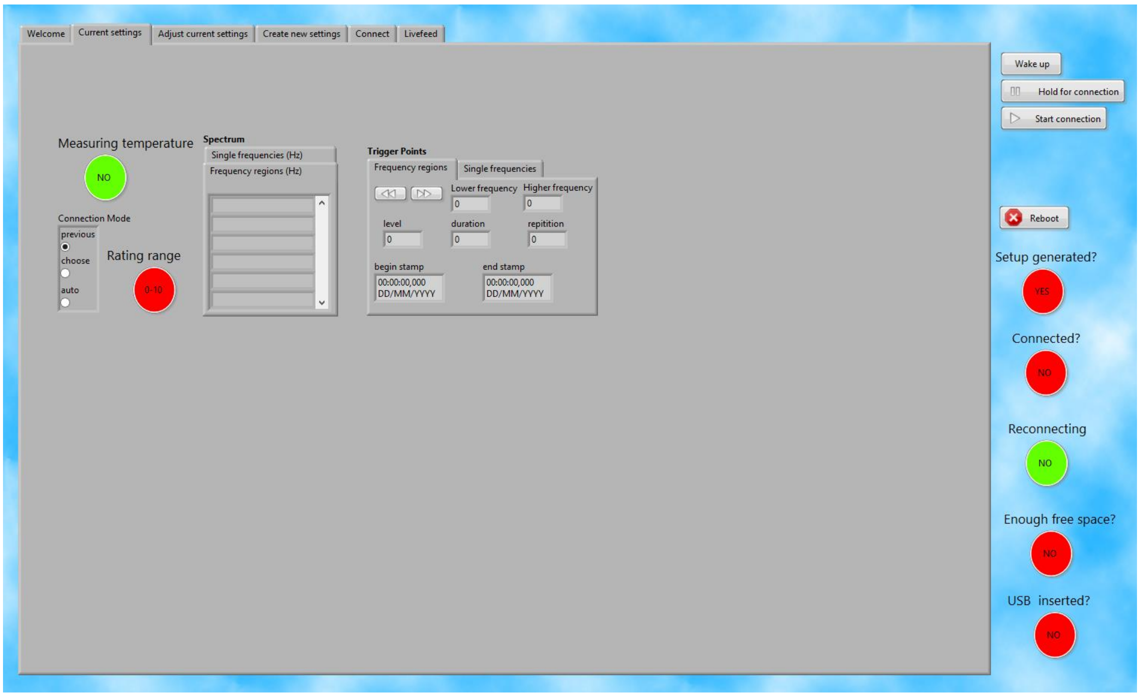

Figure 34: Deployed settings tab

6.3.4 Adjust Deployed Settings

The next tab adjust deployed settings allows the user to change the current settings of the day in question. That is, as long as they are not already deployed otherwise a system reboot is necessary before the changes take effect. This tab is displayed in Figure 35.

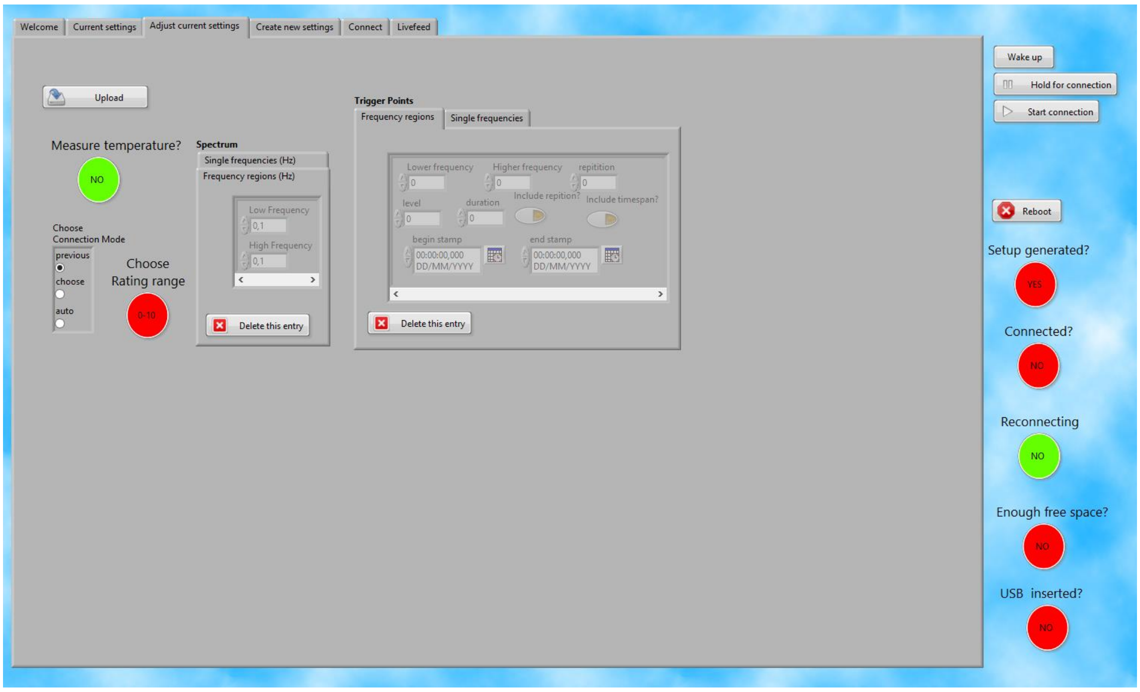

Figure 35: Adjustment tab of the deployed settings

6.3.5 Create New Settings

The previous section already explained that you could adjust the current settings. The next tab however allows you to make one from scratch and to select the day when the settings should be used.
The settings are laid out in full in 6.4 List of Settings.
All this can be seen in Figure 36.

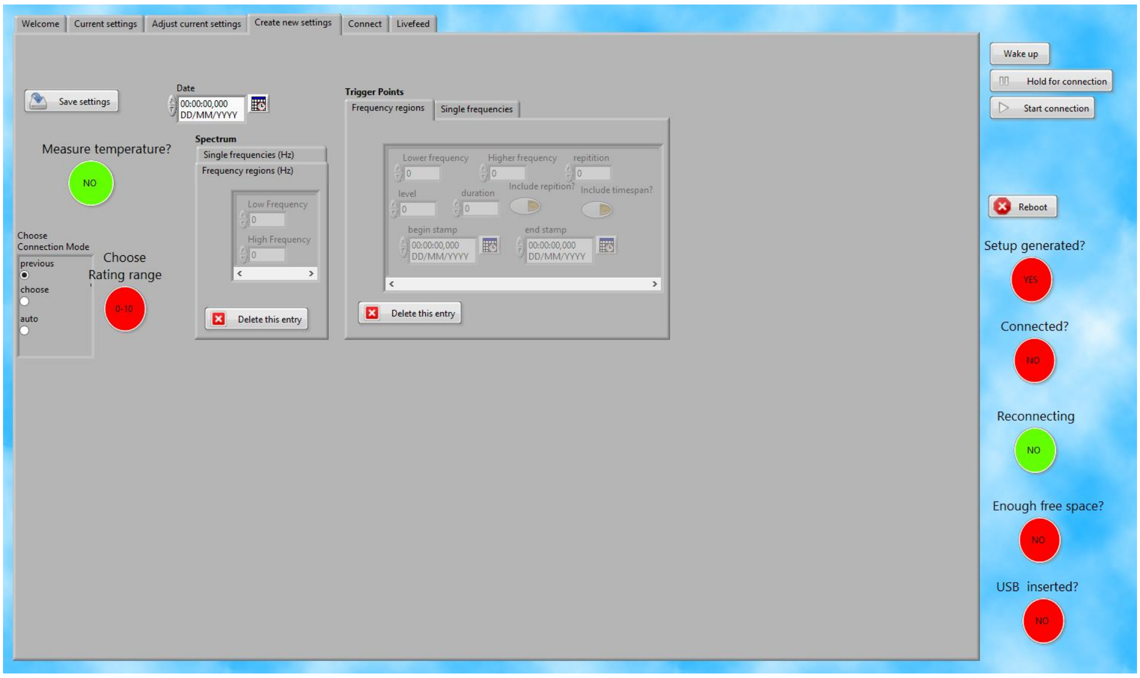

Figure 36: Tab to create entirely new settings

6.3.6 Connect

This tab shows the user errors that could have occurred. Moreover if the connection mode choose is used then available will list all the devices discovered. One can be selected and by clicking connect the myRIO will connect to the selected device. This is depicted in Figure 37.

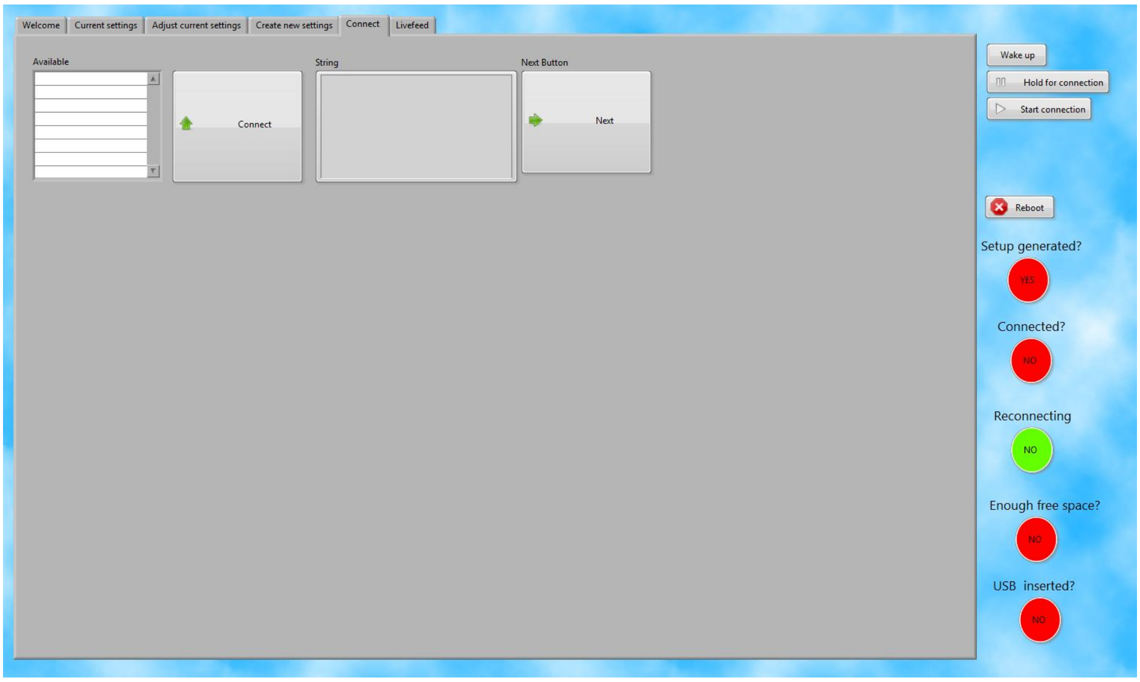

Figure 37: Connection tab

6.3.7 Livefeed

Finally the last tab allows the user to select the spectrums that are being measured. The graph at the top shows the frequency regions and the lower one the single frequencies. The displayed spectrum can be changed at the lower left by selecting another one in the list. At the lower right the user is able to see the registration coming from the patient. Of course this can be seen in Figure 38.

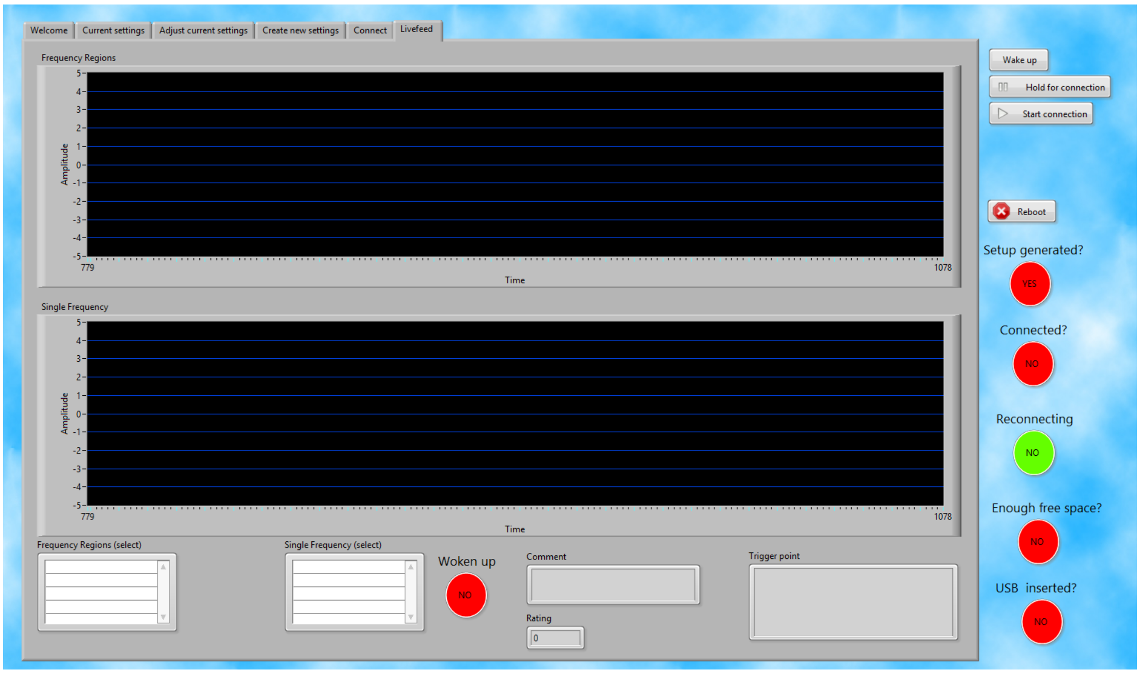

Figure 38: The Livefeed tab

6.4 List of Settings

This section will give a nice overview of all the settings that are available for the sleeper and the physician.

6.4.1 Sleeper's Settings

- Leave a comment:
  An optional field where the user can leave a comment about the awakening.

- Your experience:
  The rating from 0 to 4 or from 0 to 10 the user has to enter.

6.4.2 The Physician's Settings

- Choose connection mode:
  - Auto: myRIO searches for EEG sensors and connects to the first one it finds.
  - Choose: The user can select an EEG sensor from a list of sensors the myRIO has detected.
  - Previous: The myRIO connects to the last sensor it has successfully connected to.

- Measure temperature?:
  If set to YES the sensor will also measure the sleepers temperature.

- Choose Rating Range:
  Can be set to be between 0 and 4 or between 0 and 10.

- Spectrum Frequency regions:
  Defines the region that will be measured. Multiple regions can be stored.
  - Low Frequency: the lower frequency of the region.
  - High Frequency: the higher frequency of the region.

- Single frequencies:
  Defines a single frequency that will be measured. Multiple frequencies can be measured.

- Trigger points for frequency regions:
  Defines the conditions for an intentional awakening that tracks a frequency region.
  - Low frequency: The lower frequency of the region.
  - High frequency: The higher frequency of the region.
  - Repetition: How many times the trigger goes off if it occurs multiple times in one night.
  - Level: The minimum value the signal needs to be.
  - Duration: How long the signal needs to be above the value set in level.
  - Timespan: During which part of the night the trigger will be evaluated.
  - Include repetition?: If not set than the trigger will ignore the repetition value and will keep repeating.
  - Include timespan?: If not set than the trigger will ignore the timespan value and the whole night will be evaluated.
- Trigger points for single frequencies:
  Defines the conditions for an intentional awakening that tracks a single frequency.
  - Frequency: The frequency that is being tracked.
  - Repetition: How many times the trigger goes off if it occurs multiple times in one night.
  - Level: The minimum value the signal needs to be.
  - Duration: How long the signal needs to be above the value set in level.
  - Timespan: During which part of the night the trigger will be evaluated.
  - Include repetition?: If not set than the trigger will ignore the repetition value and will keep repeating.
  - Include timespan?: If not set than the trigger will ignore the timespan value and the whole night will be evaluated.

7 Conclusion & Future Work

This chapter concludes the project and also looks at some problems that were not resolved.
The first part draws an overall conclusion.
The following part discusses the unresolved problems.
And the last part discusses the possible future extending development of the sensor.

7.1 Conclusion

This project has successfully ended and a user friendly open-source sleep EEG measurement system has been developed.

This system is capable of processing the data without supervision. Activity in the brain is saved for later research and analysed for real-time interaction with the patient. Indeed, the system can detect certain sleep states programmatically and act upon them by waking the patient who with great ease can register a rating of the activity that just has passed.

Which data will be stored is completely up to the physician. Because through the use of the developed EEG measurement system, the physician is able to choose whatever frequency spectrums and whatever trigger points have to be saved. Furthermore, at any point in time the physician can jump in and take a look at the signals that are measured. The physician is also able to wake up the patient at any given moment.

The many different operable ways and settings of this EEG measurement system will make it a popular system for researchers of sleep EEG. The final product is depicted in the following figure, Figure 39.

Figure 39: Final product

7.2 Current Issues

7.2.1 Time and Date Setup

Although the RTC module has been mentioned and the setting of the date and time. This has not yet been implemented in the program due to late discovery of the absence of time keeping of the myRIO.

This is only requires a small vi, hence the implementation should not take long and not be difficult at all to integrate in the current code.

7.2.2 Duration of the Temperature Measurement

It should be noted that the temperature measurement takes 200 milliseconds to complete. Having a sample interval of 4 milliseconds there are 50 samples lost. Since the temperature measurement only takes place every minute this can be neglected.
However the trigger point monitoring depends on it. Indeed, during the temperature measurement the loss of samples need to be circumvented by freezing the duration timer. This is a solution, but it will certainly be better to have faster temperature sensor. Preferable one that is faster than 4ms. That way a temperature measurement is taken between two EEG samples and no EEG sample is ever lost.

7.3 Towards 19 Electrodes

Currently the sensor consists out of 3 electrodes. However the need for more electrodes has risen. This will increase the spatial resolution of the EEG measurement. It will also generate a lot more data. But this should hardly be a problem since USB devices of varying memories can be plugged into the myRIO.

7.4 Processing with a Smartphone

It has been mentioned several times that the sensor was built in such a way that almost any Bluetooth device can connect with it and use it as an EEG sensor.

Therefore a great opportunity arises to create a smartphone application that is able to do the same processing as the myRIO. Moreover, this project has already been confirmed to be a bachelors thesis in 2015-2016.

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Attachments

Attachment A: List of Components

Attachment B: Sallen-Key Low Pass Filter Calculation

Attachment C: The Second PCB

Attachment D: The Microcontroller Code

Attachment A:
List of Components

NI myRIO:

Embedded hardware device that is used to process the EEG signals.

Teensy 3.1:

The development board that is used to control the sensor.

Battery:

1000mAh polymer lithium ion.

Power Cell-LiPo Charger/Booster:

This board charges the battery and regulates the power coming from the battery to the sensor.

BlueSMiRF Silver (two units):

This is the Bluetooth modem that sends the signals to the myRIO.
One unit for the sensor and one for the myRIO.

INA114PA:

This 8-pin DIP (Dual in-line package) IC is the differential amplifier also called an instrumentation amplifier.

TLC277CP (two units):

8-pin DIP IC that contains two operational amplifiers.

Real-Time Clock Module:

Module that keeps track of date and time without supply voltage by using a backup battery.

Buzzer:

Can be any kind of buzzer.

myRIO MXP protoboard:

This protoboard can connect to the MXP port of the myRIO and used to solder components on.

DS18B20:
The digital temperature sensor.

BC546 (two units):
The NPN transistor.

BC556 (two units):
The PNP transistor.

Capacitors:

Value	Amount
10µF	2
10pF	1
100pF	2
1µF	2
0.68µF	4
0.1µF	2
1nF	2

Resistors:

Value	Amount
2.2kΩ	9
6.8kΩ	2
1MΩ	2
100Ω	2
82kΩ	2
15kΩ	2
150Ω	2
4.7kΩ	1
200kΩ	1
10k	1

Attachment B:
Sallen-Key Low Pass Filter calculation

The general equations for the cut off frequency and quality factor of a Sallen-Key low pass filter are these[21]:

$$fc = \frac{1}{2\pi * \sqrt{R1 * R2 * C1 * C2}}, (A.1)$$

$$Q = \frac{\sqrt{R1 * R2 * C1 * C2}}{R1 * C1 + R2 * C1 + R1 * C2 * (1 - K)} (A.2)$$

To make the calculation of the values easier the capacitors will be set to the same value and the resistors will be set as a ratio[21]. Thus,

$$R1 = m * R \ (A.3)$$

$$R2 = R (A.4)$$

$$C1 = C2 = C (A.5)$$

Substituting this into the general equation (A.1) and (A.2) leads to:

$$fc = \frac{1}{2\pi * R * C * \sqrt{m}} (A.6)$$

$$Q = \frac{\sqrt{m}}{1 + 2 * m - m * K} (A.7)$$

In order to have a stable low pass filter Q should always be greater than zero. Furthermore, K has to be 100. The nominator of (A.7) will be greater than zero for whatever value of m or K. Thus,

$$\begin{aligned} &Q > 0 \\ &1 + 2 * m - m * K > 0 \\ &1 + 2 * m - m * 100 > 0 \\ &1 - 98 * m > 0 \\ &1 > 98 * m \\ &\frac{1}{98} > m \end{aligned}$$

$$0.01 > m$$

Notice that the factor K in (A.7) has a negative sign. Therefore when K becomes large Q gets negative real quick. As a counteraction m should be taken very small, hence the value chosen will be m=0.001. The resistor values that were chosen are 100Ω for R_2 and 100kΩ for R_1. The required cut off frequency is 80Hz. Substituting these values into equation (A.6) and solving for C gives:

$$fc = \frac{1}{2\pi * R * C * \sqrt{m}}$$
$$C = \frac{1}{2 * \pi * f_c * R * \sqrt{m}}$$
$$C = \frac{1}{2 * \pi * 80 * 100 * 10^3 * \sqrt{0.001}}$$
$$C = 0.62\ \mu F$$

The available capacitor value closest to 0.62μF is 0.68μF. The R_2 value is also adjust to 82kΩ. Calculating backwards with the following eventual values:

$$C = 0.68\mu F$$
$$R_1 = 100\Omega$$
$$R_2 = 82k\Omega$$
$$K = 100$$

leads to (with (A.3 and A.4):

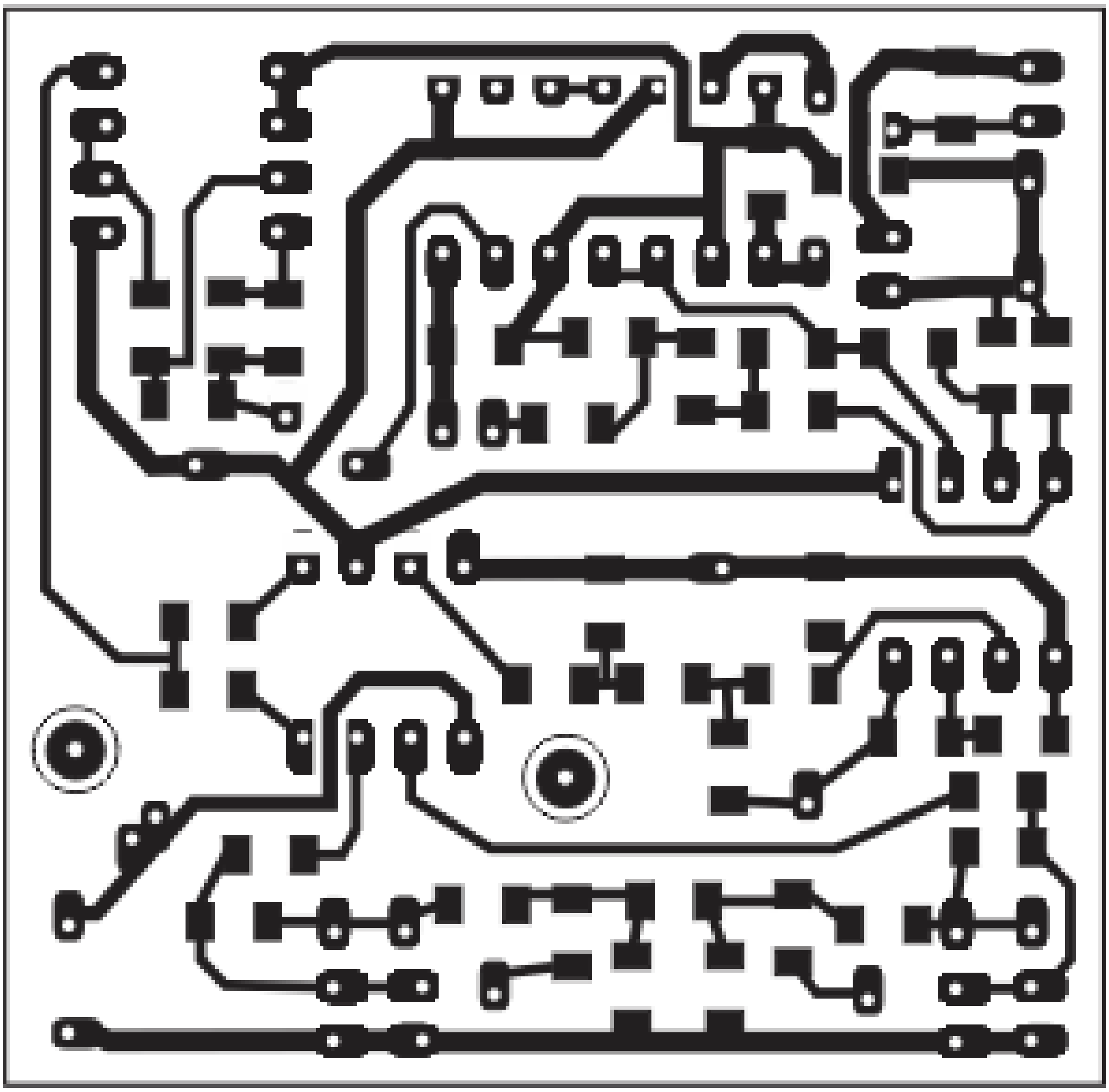
$$fc = \frac{1}{2\pi * R * C * \sqrt{m}}$$
$$fc = \frac{1}{2\pi * 82 * 10^3 * 0.68 * 10^{-6} * \sqrt{0.0012}}$$
$$fc = 82{,}39Hz$$

And:

$$Q = \frac{\sqrt{m}}{1 + 2 * m - m * K}$$
$$Q = \frac{\sqrt{0.0012}}{1 + 2 * 0.0012 - 0.0012 * 100}$$
$$Q = 0.039$$

Since Q is greater than zero and the cutoff frequency is slightly larger than 80Hz, these values will do fine.

Attachment C: The Second PCB



A keen reader of course notices that the IC on the top level in the middle is not discussed anywhere in this thesis. This was meant for the analogue 50Hz LMF90 filter but has not been implemented. It would have doubled the cost of the EEG sensor. Though, it could still be implemented in the future.

Attachment D:
The Microcontroller Code

finalversion §

```
#include <OneWire.h>
#include <DallasTemperature.h>
#include <TimedAction.h>

#define ONE_WIRE_BUS 12
#define HWSERIAL Serial1
#define EEGsignal A9

boolean tempmode = true;
int lastConnect= 0;
char highval;
char lowval;
char mc4;
char mc3;
char mc2;
char mc1;
char EEGCode = B01101101;
char tempCode = B01111010;
unsigned char tempAddress[8];
String connectedMac;


OneWire oneWire(ONE_WIRE_BUS);
DallasTemperature sensors(&oneWire);
TimedAction EEGsample = TimedAction(10000,4,sendEEG);
TimedAction Tempsample = TimedAction(10000,1000,sendTemp);


void setup()
{
  analogReadResolution(13);
  Serial.begin(115200);
  HWSERIAL.begin(115200);
  pinMode(EEGsignal, INPUT);
  sensors.begin();
  sensors.getAddress(tempAddress,0);
  makeConnection();
}
```

```
void loop()
{
     EEGsample.check();
  if(tempmode)
  {
     Tempsample.check();
  }

  if(HWSERIAL.available()>0)
  {
    char incoming = HWSERIAL.read();
    if(incoming =='T')
    {
      sendTemp();
    }
    else if(incoming == 'S')
    {
      lastConnect = millis();
    }
    else if(incoming == 'X')
    {
      breakConnection();
      makeConnection();
    }
  }
  else if(millis()-lastConnect>20000)
   {
     Serial.println("break");
     breakConnection();
     reconnect();
   }
}
```

```
void sendEEG()//take the EEG signal
{
  int EEGvalue = analogRead(EEGsignal);
  formatmillis();
  format(EEGvalue);
  sendcode(EEGCode);
  clearvariables();
}
void sendTemp()//take the temperature value
{
  sensors.requestTemperatures();
  int tempvalue = (int)(sensors.getTempC(tempAddress)*100);
  formatmillis();
  format(tempvalue);
  sendcode(tempCode);
  clearvariables();
}

void format(int toform)//build up the characters to send
{
  int highnum = toform/100;
  highval = byte(highnum) | B10000000;
  int lownum =toform-highnum*100;
  lowval = byte(lownum) | B10000000;
}

void breakConnection()// breaks the connection
{
  delay(100);
  HWSERIAL.print("$");  // Print three times individually
  HWSERIAL.print("$");
  HWSERIAL.print("$");
  delay(100);
  HWSERIAL.println("K,");//sends err when already disconnected
 // and does not leave command mode
  delay(100);
  HWSERIAL.println("---");//is needed to go out of CMD when module
  //is not connected en thus still in CMD when K, is used
}

```

```
void reconnect()
{
  boolean connectionmade=false;
  String taskString = "";
  char taskChar;
  while(!connectionmade)// connection is issued by MyRIO
  {
    delay(100);

    while(HWSERIAL.available()>0)
        {
          taskChar = (char)HWSERIAL.read();
          taskString += taskChar;
        }
    if(taskString.length() == 26)
    {
      if(taskString.substring(0,11) == "ESC%CONNECT")
      {
        if(taskString.substring(12,24) == connectedMac)
        {
          connectionmade =true;
          lastConnect = millis();
        }
        else
        {
          HWSERIAL.print("this device is currently coupled with ");
          HWSERIAL.println(connectedMac);
          breakConnection();
        }
      }
    }
  taskString = "";
  }
}
```

```
void formatmillis()
{
  Serial.println("m");
  int t1 = millis();
  int m4 = t1/1000000;
  Serial.println(m4);
  int t2 = t1-(m4*1000000);
  int m3 = t2/10000;
  Serial.println(m3);
  int t3 = t2- (m3*10000);
  int m2 = t3/100;
  Serial.println(m2);
  int m1 = t3-(m2*100);
  Serial.println(m1);
  mc4=byte(m4) | B10000000;
  mc3=byte(m3) | B10000000;
  mc2=byte(m2) | B10000000;
  mc1=byte(m1) | B10000000;
}

void sendcode(char type)
{
  HWSERIAL.print(type);
  HWSERIAL.print(highval);
  HWSERIAL.print(lowval);
  HWSERIAL.print(mc4);
  HWSERIAL.print(mc3);
  HWSERIAL.print(mc2);
  HWSERIAL.print(mc1);
  HWSERIAL.print('\n');
}

void clearvariables()
{
  highval=NULL;
  lowval=NULL;
  mc4=NULL;
  mc3=NULL;
  mc2=NULL;
  mc1=NULL;
}
```

```
void makeConnection()
{
  boolean connectionmade=false;
  String taskString = "";
  char taskChar;
  while(!connectionmade)
  {

    while(HWSERIAL.available()>0)
      {
        taskChar = (char)HWSERIAL.read();
        taskString += taskChar;
      }
      delay(100);
    if(taskString.length()==26)
    {
      if(taskString.substring(0,11) == "ESC%CONNECT")
        {
          HWSERIAL.println("EEGD");
          //to connect to myRIO all EEG sensors
          //(not only this one) need to identify afterwards
          connectedMac=taskString.substring(12,24);
          taskString = "";
          int setuptime=millis();
          while(millis()-setuptime<5000)
          {
            while(HWSERIAL.available()>0)
            {
              taskChar = (char)HWSERIAL.read();
              taskString += taskChar;
            }
            delay(100);
            if(taskString == "TC")
            {
              tempmode =true;
              sensors.setResolution(10);//10 bits ->0.25
              connectionmade= true;
              lastConnect=millis();
              HWSERIAL.print("scmsg");
              Serial.println("cont");
            }
```

```
            else if(taskString == "TA")
            {
              tempmode = false;
              sensors.setResolution(12);//12bits->0.0625, 11bits->0.125
              connectionmade= true;
              lastConnect=millis();
              HWSERIAL.println("scmsg");
              Serial.println("ask");
            }
            else if(taskString.length()!=0){
              Serial.println(taskString);
              HWSERIAL.println("camsg");
            }
            taskString= "";
          }
          if(!connectionmade)
          {
            HWSERIAL.println("sferr");
            breakConnection();
          }
        }
      }
      taskString = "";//reset taskString
    }
}
```

Auteursrechtelijke overeenkomst

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EEG signaal analyse

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Jaar: **2015**

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Datum: **1/06/2015**