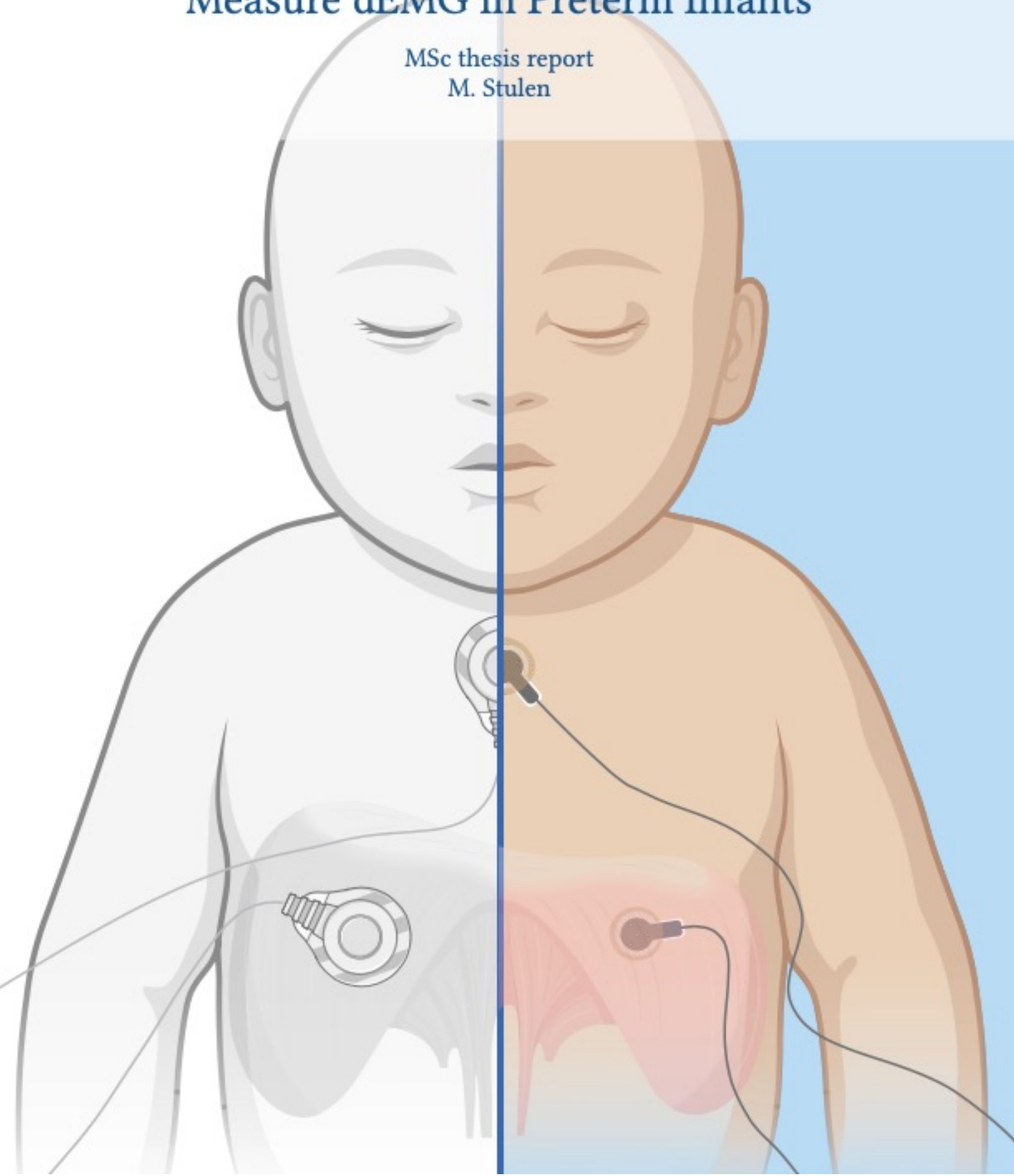


Creating a Thin, Flexible Biopotential Electrode to Measure dEMG in Preterm Infants

MSc thesis report
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MSc Thesis Report

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Abstract—Introduction High impact respiratory conditions are common amongst preterm infants. Measuring the electromyogram of the diaphragm (dEMG) is a very promising technique to accurately measure respiratory state. However, the currently used Ag/AgCl electrodes are reported to be big and thick, making them less suitable to measure dEMG in preterm infants. Therefore, the aim of this study was to create a prototype for a thin, flexible electrode that could measure dEMG in preterm infants with at least the same signal quality as the currently used Ag/AgCl electrode.

Methods To do so, design requirements and wishes were set. Literature and three tests were used to assess whether a certain prototype met the design requirements. If one of the design requirements was not met, an iteration cycle was started and the prototype was redesigned. With the final version of the prototype, a proof of principle test was performed, where dEMG measurements were conducted on a healthy female adult using the prototype and the standard Ag/AgCl electrodes simultaneously. **Results** The final design was created from two layers of Shieldit conductive fabric ironed onto cotton and one insulating layer of TPU. These layers were connected to a shielded cable by weaving the copper wire through the fabric.

Ultimately, not all design requirements were met. The frequency plot of the final prototype still showed a peak at 50 Hz, indicating insufficient shielding from electromagnetic interference. However, the final prototype was indeed dry, thinner, and more flexible than the Ag/AgCl electrode. The RMSE value for the prototype (0,5984 mV) was smaller than that for the Ag/AgCl electrode (0,5998 mV), although the opposite was true for the SD (prototype: 0,1031 mV, Ag/AgCl: 0,0316 mV). In addition, it was proven that the final prototype could be used to measure dEMG in healthy adults. The breathing frequency measured by the final prototype was equal to the breathing frequency measured by the Ag/AgCl electrode, whilst showing no significant difference in amplitude of the peaks of the breathing curve ($p = 0,0830$).

Discussion Design improvements could be made by eliminating the 50 Hz peak, decreasing diameter even further, creating a new version where the cable could pivot around the electrode or exploring wireless options.

Conclusion In this paper, it has been proven that a textile prototype is well capable of measuring dEMG in humans and offers several improvements over the Ag/AgCl electrode.

Index Terms—biopotential electrodes, textile electrode, electrode material, dEMG, electromyogram, diaphragm, preterm infants, neonates.

I. INTRODUCTION

A. Rationale

ACCORDING to the World Health Organization (WHO) a birth is defined to be preterm when a baby is born alive before the first 37 weeks of a pregnancy are completed [22]. These infants are born immature and their lungs are underdeveloped. Therefore, respiratory diseases, such as respiratory distress syndrome (RDS) and apnea of prematurity (AOP), are common in these patients [15]. Within extremely preterm infants (gestational age (GA) < 28 weeks [22]) 80% suffers from RDS [23, 30] and in almost all infants that were born before a GA of 34 weeks AOP occurs [9, 15]. These diseases have high impact since RDS is a major contributor to neonatal mortality worldwide [12] and AOP can cause neural morbidity and neurological development impairment [15]. Therefore, continuous respiratory monitoring at the neonatal intensive care unit (NICU) is essential.

At the NICU, various techniques to monitor respiratory state already exist. Flow and pressure measurements, chest impedance (CI) measurements and respiratory inductance plethysmography (RIP) can be performed [15]. However, these techniques have downsides. Flow and pressure measurements are sensitive to air leakage, especially when providing non-invasive respiratory support. Chest impedance measurements can easily be disturbed by body movements and clinical handling, they do not provide any information on breathing effort, nor can be used to classify apnea. In addition, RIP measures are highly sensitive to displacement of the bands that are used. Lastly, all techniques mentioned above measure the secondary effects of respiratory muscle activity, resulting in a delayed detection of specific breathing patterns.

To enhance respiration monitoring, a novel technique has been developed to measure the activity of the diaphragm, the most important and first-activated muscle during a breathing cycle. Therefore, measuring diaphragm activity using electromyography (dEMG) is a very promising technique that has potential to measure work of breathing, detect and classify apnea and to synchronize ventilation with the natural breathing frequency of the patient [15].

Currently, dEMG can be measured using invasive as well as non-invasive techniques. Within preterm infants however, non-invasive methods are preferred. When performing non-invasive dEMG measurements, sensor belts or biopotential surface electrodes can be used. A sensor belt, such as the Bambi Belt, uses dry metal electrodes that are pressed to the

skin using a silicone band around the thorax of the infant [5]. Unfortunately, the band comprises a relatively large area of the baby's chest, the measurements are sensitive to band displacement and the band cannot be easily reused. Therefore, an approach using biopotential surface electrodes is preferred. When performing dEMG measurements using biopotential surface electrodes, three standard electrocardiogram (ECG) Ag/AgCl electrodes are placed on the thorax of the infant: one reference electrode on the sternum and two active electrodes in the mid-clavicular line (Figure 1) [15]. This type of electrode uses electrolytic gel to enhance contact between electrode and skin, hence they are called 'wet' electrodes. The electrolytic gel improves contact between skin and electrode, lowering skin-electrode impedance and enhancing the quality of the biopotential signal [3].

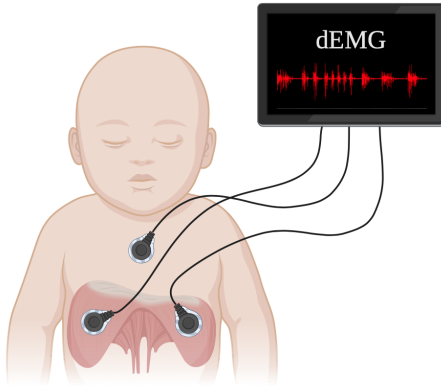


Figure 1: A schematic drawing of the placement of electrodes on an infant. One electrode is placed on the sternum and two electrodes placed in the mid-clavicular line, one left and one right. Currently, Ag/AgCl electrodes (here, drawn as round Ag/AgCl electrodes) are used to measure the electromyogram of the diaphragm (dEMG).

Nevertheless, these electrodes have downsides as well. The signal they generate remains sensitive to artifacts and the gel they use can dehydrate, which makes them unsuitable for long-term monitoring [15]. Most importantly however, the snap connection between cable and electrode is thick and hard, which can be uncomfortable for the patient and which causes guardians to be reluctant to participate in dEMG research. Therefore, to integrate dEMG as a novel technique to continuously monitor respiratory state in preterm infants at the NICU, a new biopotential surface electrode is needed.

B. Research goal

In this paper, it was aimed to develop a prototype electrode for measuring dEMG in preterm infants, which is dry, thinner and more flexible than the currently used Ag/AgCl electrode, whilst maintaining the same signal quality.

II. METHODS

A. Current measurement system

During breathing, activity of the diaphragm generates a biopotential signal. This signal is transmitted from skin, through the electrolytic gel, to the silver sensing part of the

Ag/AgCl electrode. In this paper, the rectangular Kendall Ag/AgCl ECG electrode (CardinalHealth, Dublin, Ireland [14], Figure 2) was considered as the standard Ag/AgCl electrode, to which the prototypes were compared.

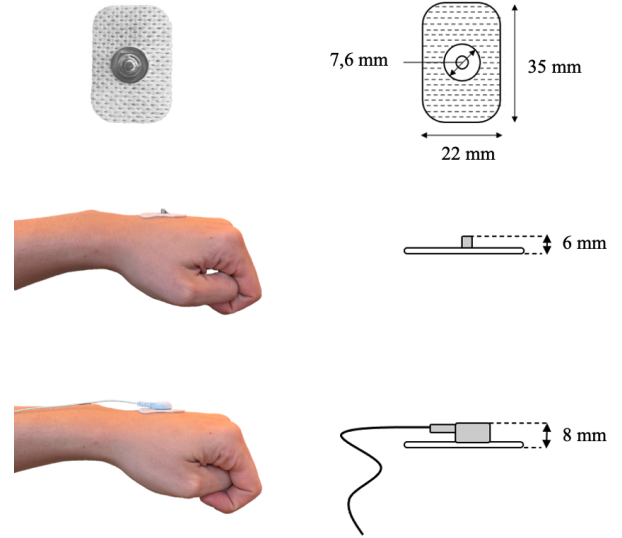


Figure 2: Dimensions of the currently used rectangular Kendall Ag/AgCl cloth electrode. The adhesive part of the electrode is 22 x 35 mm in size. The silver sensing part has a diameter of 7,6 mm and a thickness (including cloth) of 6 mm. The snap connector is attached to the silver sensing part, creating a total thickness of 8 mm.

After the biopotential signal is sensed by the electrode, the signal is transmitted to a shielded cable via a snap connector. The cable is connected to the Diphia-16 amplifier (DEMCON Macawi Respiratory Systems, Best, the Netherlands [10]). In this orange smartphone-sized battery powered amplifier, the data acquisition, amplification and preprocessing takes place. The Diphia-16 measures the biopotential signal with a sampling frequency of 500 Hz, after which a low-pass filter is used.

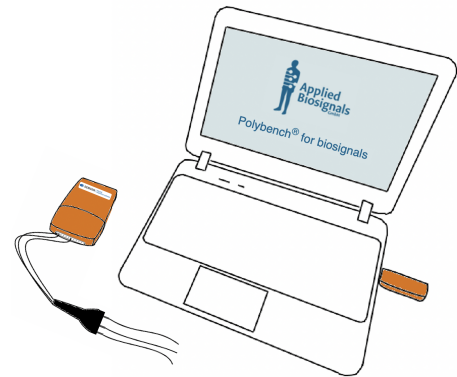


Figure 3: Schematic drawing of a measurement set-up using the Diphia-16. Three cables (black) go to the header of the Diphia-16 amplifier (orange, on the left). Here, pre-processing takes place, after which the signal is transmitted to the USB dongle (orange, right side of the laptop). The laptop uses the Polybench application to visualize the measured biopotential signals.

Therefore, signals with a frequency up to 160 Hz can be measured. The amplifier that is used, is a 24-bit amplifier with a resolution of 65 nV/bit and an input impedance of 2 GΩ. [10] After conversion to a digital signal, the data is transmitted to a transceiver USB dongle. This orange USB dongle is connected

to a laptop, on which the Polybench application can be used to visualize the measured signal (Figure 3).

B. Design process

The design process of the prototype was divided into various phases (Figure 4): setting design requirements and wishes, designing a first prototype that enters various iteration cycles and a phase where the final design is used in a proof of principle test.

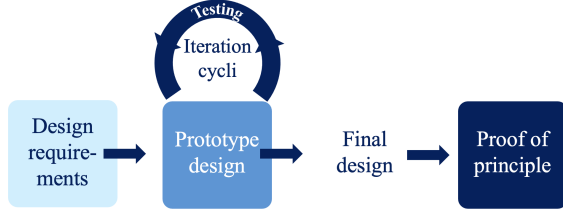


Figure 4: Flow diagram of the design process

C. Design requirements

First, in association with the NICU at the Amsterdam UMC (Amsterdam, Netherlands) and DEMCON Macawi Respiratory Systems (Best, Netherlands) the design requirements and wishes for the electrode were set and a table showing the design requirements and wishes was created.

D. Prototype design

A first version of the prototype was created based on literature. With this version, the iteration phase was started and it was analyzed how well this version met the design requirements. Some of the requirements were evaluated using literature, others were assessed by using one of three tests, of which their methods were included in Appendix B:

- Test A: assessing electrical resistance of the cable and resistance against pulling forces.
- Test B: measuring skin-electrode impedance.
- Test C: quantifying the root mean square error (RMSE) value and the standard deviation (SD) of the measured signal and assessing the frequency plot.

If a prototype would not meet at least one of the design requirements, an iteration cycle was started and the prototype was adjusted. Finally, the version of the prototype that met most design requirements after three months of iterations, was considered the final design and its performance was assessed during a proof of principle test.

E. Proof of principle test

In this test, the final prototype was used to measure dEMG on a healthy subject. Two laptops running the Polybench application, two DPHA-16 amplifiers (DEMCON Macawi Respiratory Systems, Best, Netherlands) and two USB dongles were used. One laptop, DPHA-16 and its USB dongle were connected to the standard Ag/AgCl electrodes. The other laptop and DPHA-16 were connected to the final prototype. Subsequently, the electrodes were placed as shown in Figure

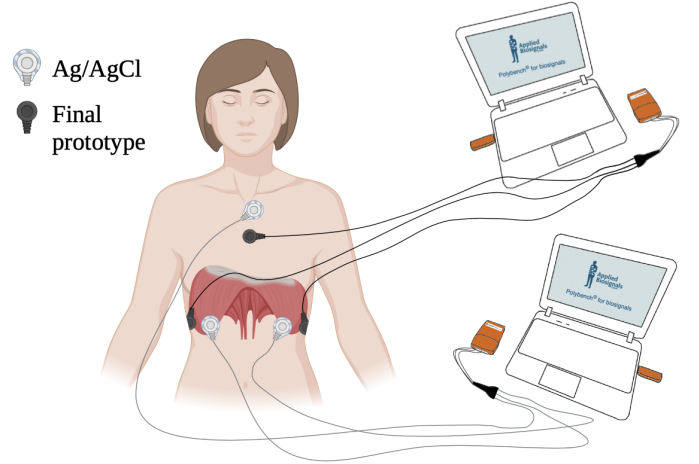


Figure 5: Placement of electrodes during the proof of principle test. In white the standard Ag/AgCl electrode and in black the final prototype.

5, so dEMG measurements with the two different electrodes could be performed simultaneously.

At the start of the measurement, a metronome was set with a frequency of 60/min. The subject was asked to breathe in and out on the rhythm of the metronome, resulting in a breathing frequency of 30/min. Raw EMG data was measured, which was recorded in Polybench as .Poly5 file. Finally, an (adjusted) Matlab script [26, 16] was used to post-process the data and to extract a breathing curve.

First, this Matlab script subtracted lead 1 from lead 2, added a high pass filter to remove offset and drift and plotted the ECG signal that followed. Then, the ECG peaks were removed and filled with a copy of previous data and a 50 Hz notch filter. Lastly, a breathing curve was extracted. The full signal was included in Appendix F. The performance of the prototype was compared to the Ag/AgCl electrode by plotting the breathing curves and comparing measured breathing frequencies and peak amplitudes (using a paired t-test). In addition, a Harris profile for both electrodes was created to compare how well both electrodes met the design requirements and wishes.

Since the subject was the author herself and no other participants were involved in the proof of principle test, it was concluded that this test was no subject to WMO regulations, of which a statement was included in Appendix A.

III. RESULTS

Figure 6 shows the total design process. First, the design requirements and wishes were set. A first prototype was created and after two iterations, a final design was used in a proof of principle test.

A. Design requirements

The design requirements and wishes were divided into categories corresponding to different parts: the electrode itself, the cable-electrode connection, cable and skin-electrode interface requirements (as shown in Table 1). The fifth column in Table 1 shows what method was used to verify whether a prototype met a certain requirement or wish. If a prototype would not meet at least one of the design requirements, an

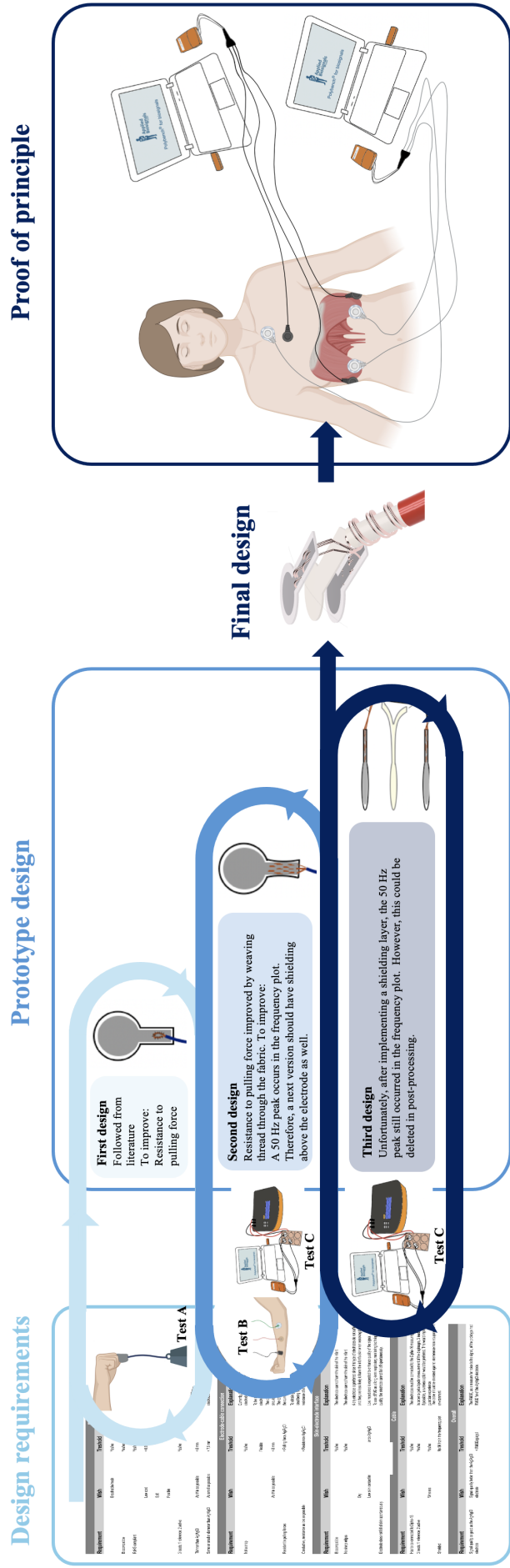


Figure 6: Design process. First design requirements were set and a first design was created. To verify whether a design would meet the design requirements various tests were performed. In the end, three iteration cycles were needed to come to a final design, which was used in a proof of principle test.

Electrode				
Requirement	Wish	Threshold	Explanation	Test
Biocompatible RoHS compliant	Breathable/mesh	Yes/no	A material is preferred that is permeable to moisture and air, whilst its properties are not influenced by the high humidity in the incubator.	Literature
		Yes/no	The electrode cannot harm the skin of the infant	Literature
		Yes/no	No lead, cadmium, mercury, chromium, PBB, PBDE, DEHP, BBP, DBP, DIBP may be used.	Literature
	Low cost	< €0,68	The currently used Ag/AgCl electrode costs €0,68 per electrode. The newly designed prototype preferably costs less than the current situation. After this holds: the lower the costs, the better.	Literature
	Soft		Preferably, the material is soft to the infant's skin.	Observation
	Flexible		Preferably, the material is flexible, so skin conformance is increased.	Observation
Thinner than Ag/AgCl	As thin as possible	< 8 mm	The thickness of the electrode should be smaller than the currently used Ag/AgCl electrode. After this requirement is met, the following holds: the thinner the better.	Caliper
Same or smaller diameter than Ag/AgCl	As small as possible	< 7,6 mm	The diameter of the electrode should be smaller than the currently used Ag/AgCl electrode. After this requirement is met, the following holds: the smaller the better.	Caliper
Electrode-cable connection				
Requirement	Wish	Threshold	Explanation	Test
Not a snap		Yes/no	Currently, the snap connection is the biggest problem regarding thickness of the electrode. Hence, an electrode that does not use a snap connection is a must.	Observation
	Flexible	Yes/no	To be able to move the cable in various directions without putting too much force on the electrode or the electrode-cable connection, the connection is preferred to be flexible.	Observation
Thinner than Ag/AgCl	As thin as possible	< 8 mm	The currently used snap connection has a thickness of 8 mm. The connection from electrode to cable should be equally thin or thinner than the currently used cable.	Caliper
Resistant to pulling forces		< Pulling force Ag/AgCl	The connection between electrode and cable cannot break faster in the prototype than in the current situation, when the cable is pulled.	Test A
Conductive, resistance as low as possible		< Resistance Ag/AgCl	To obtain good signal, the resistance of the connection should be as low as possible, ideally the resistance of a plain copper wire. The threshold was set to the electrical resistance of the currently used Ag/AgCl electrode.	Test A
Skin-electrode interface				
Requirement	Wish	Threshold	Explanation	Test
Biocompatible		Yes/no	The electrode cannot harm the skin of the infant	Literature
No sharp edges		Yes/no	The electrode cannot harm the skin of the infant	Observation
	Dry	Yes/no	A dry electrode is preferred, since this type of electrodes do not suffer from gel dehydration and they are less likely to harm the infant's skin when removing the electrode.	Observation
	Low skin contact impedance	< impedance Ag/AgCl	Low impedance is expected to enhance quality of the signal.	Test B
Electrode does not fall of skin spontaneously		Yes/no	To use dEMG as a long-term respiratory monitoring technique and obtain good signal quality, the electrode cannot fall off spontaneously.	Observation
Cable				
Requirement	Wish	Threshold	Explanation	Test
Has to be connected to Diphera-16 3 leads: 1 reference, 2 active		Yes/no	The electrodes must be connected to the Diphera-16 measuring device.	Observation
		Yes/no	In order to get a bipolar measurement of the diaphragm, 3 leads are needed.	Observation
	Wireless	Yes/no	If possible, a wireless option would be preferred. This would improve clinical handling and guardian experience.	Observation
Shielded		No 50 Hz in the frequency plot	The cable should be shielded against interference from equipment in the direct environment.	Test C
Overall				
Requirement	Wish	Threshold	Explanation	Test
Signal quality as good as the Ag/AgCl electrode	Signal quality better than the Ag/AgCl electrode	< RMSEAg/AgCl	The RMSE of the prototype had to be smaller than the RMSE for of the Ag/AgCl electrode.	Test C
Signal quality as good as the Ag/AgCl electrode	Signal quality better than the Ag/AgCl electrode	< SDAg/AgCl	The SD of the prototype had to be smaller than the SD for of the Ag/AgCl electrode.	Test C

Table 1: Design requirements of a new electrode, organized by electrode, electrode-cable connection, skin-electrode interface, cable and overall requirements. The requirements were distinguished from wishes. A threshold for each requirement or wish was set along with a testing method to define whether the specific requirement or wish was met. Lastly, a column containing a short explanation regarding a certain requirement or wish was added.

iteration cycle was started and the prototype was adjusted. Wishes were viewed as a bonus during the design process, not a must.

Electrode:

Literature was used to define whether a certain material was biocompatible and Restriction of Hazardous Substances Directive (RoHS) compliant [24]. The costs of the material followed from literature and would preferably be lower than the buying price for the Ag/AgCl electrode (€0,68 [14] per electrode).

Electrode thickness had to be less than 8 millimeters and was preferred to be as thin as possible. The diameter of the electrode had to be below 7,6 mm for which the same held: the smaller the diameter, the better. Thickness and diameter of the electrode were measured using a caliper.

Electrode-cable connection:

The connection that was used to connect the electrode to a cable was not allowed to be a snap connection and had to be thinner than the currently used snap connection (8 mm), which was measured using a caliper.

The connection between cable and electrode had to be resistant to pulling forces, so the cable will not break when it is pulled during clinical handling. When pulling the cable, the resistance to breaking had to be as high as or higher than the current connection between cable and Ag/AgCl electrode. In addition, the connection had to be conductive to electricity. Ideally, the connection would have the electrical resistance of a continuous copper wire. However, a resistance lower than the resistance of an Ag/AgCl electrode and a snap connection was acceptable. The resistance to pulling forces and electrical resistance were measured in Test A.

Skin-electrode interface:

Biocompatibility of the materials touching the skin was assessed by searching literature. No sharp edges were allowed in the design.

The prototype would preferably be a dry electrode. Normally, the gel in an Ag/AgCl electrode secures the electrode to the skin and lowers the skin-electrode impedance [3]. In a dry electrode the gel is absent, so a different method to secure the prototype to the skin is needed. In addition, the influence on skin-electrode impedance needed to be analyzed. Therefore, skin-electrode impedance for various adhesion methods was measured in a second test, Test B.

Cable:

The prototype had to be connected to the Dipha-16 amplifier, for which an adjusted amplifier header and pin headers were created. In addition, a bipolar lead configuration was used. Therefore, a certain version of the prototype had to be manufactured in threefold: one for each lead. However, if possible, a wireless option would have been preferred.

Since the configuration would be used in an intensive care environment, the electrodes had to be connected to shielded cables to minimize electromagnetic interference. To test whether the configuration would not pick up any electromagnetic frequencies from the environment and to assess the quality of the signal, a last test (Test C) was performed. When analyzing the frequency plots following from this test, a 50 Hz peak caused by the mains electricity was not allowed to occur.

Overall:

The goal of this paper was to create a thin, flexible electrode that could measure dEMG with at least the same signal quality as the Ag/AgCl electrode. To compare signal quality, RMSE and SD values were calculated in Test C. The RMSE value as well as the SD of the final prototype had to be equal or smaller than the values for the Ag/AgCl electrode.

B. Iteration cycles

A literature study showed that textile or polymer based electrodes were most suitable to create a new thin, flexible electrode to measure dEMG in preterm infants (Appendix D). A polymer electrode made from TPU and silver conductive ink and a textile electrode from cotton and Shieldit conductive fabric [28] were created (Appendix D). In the end, the textile option was found to be most suitable. In this previous literature [7], a conductive textile electrode integrated in a smart jacket was connected to its cable by sewing the cable on the fabric using conductive thread. This was used as an inspiration to create the first design (Appendix D).

It was tested how well this design met the design requirements. From Test A followed that the electrode-cable connection could not resist sufficient pulling forces, so a different connection that could resist higher forces was needed. This started the first iteration cycle (Appendix D) towards a second design. In the second design (Appendix E), a different electrode-cable connection was used. In addition, various methods to

attach the electrode to the skin were evaluated by performing impedance measurements in Test B and by comparing signal quality measures in Test C (Appendix E). It followed that the best signal followed from using a band-aid to attach the electrode to the skin. However, frequency plots showed that the electrode was still sensitive to electromagnetic influences from the environment even though a shielded cable was used (Appendix E). Therefore, in the final design it was aimed to shield the electrode itself as well.

C. Final design

The inside of a passive shielded cable consists of the copper wire of the cable, a PVC insulation (white), a spiral shield and a PVC jacket (black) [2]. The spiral shield creates a Faraday's cage, so everything within the cage (the copper wire from the cable) is protected from electromagnetic interference. To create a prototype where the electrode itself was shielded as well, the design as shown in Figure 7 was created.

First, the Dipha-16 header was adjusted so it could be connected to a shielded prototype (Figure 8). The cables originating from a standard 3 lead Dipha-16 head were formed into a black female connector, in which pin headers could be inserted. Every electrode was connected to two pin headers: one for shielding and one for the inner cable.

To create the electrode, two pieces of Shieldit conductive fabric [28] were ironed onto two separate pieces of cotton. Then, the spiral shield around the cable was split from the cable itself. The copper wire of the cable was connected to one piece of Shieldit/cotton and the spiral shield was woven into another piece of Shieldit/cotton. Then, the two pieces of fabric were placed back-to-back against each other with an insulating layer of TPU between them. Finally, the cable connection was covered using heat shrink tube. The full manufacturing process was included in Appendix F. An image of the final prototype and its dimensions relative to the Ag/AgCl electrode could be found in Figure 9.

When the second version of the prototype was assessed, various methods to adhere the electrode to the skin were analyzed (Appendix E). First, impedance measurements (Test B) were performed. It was found that the Ag/AgCl electrode had an average skin-contact impedance of 85,6 k Ω (95% confidence interval: 69,6 - 101,6 k Ω) prototype band aid 247,9 k Ω (95% confidence interval: -46,1 - 541,9 k Ω). However, due to the high input impedance of the Dipha-16 amplifier (2G Ω), these high impedance values did not influence the quality of the signal as shown in Test C (Appendix E). From this test, it followed that the best way to adhere the electrode to the skin was by using a band-aid. Therefore, the final design as shown in Figure 7 was adhered to the skin using a band-aid.

D. Assessment of the prototype

First, it was analyzed whether the final design met the design requirements.

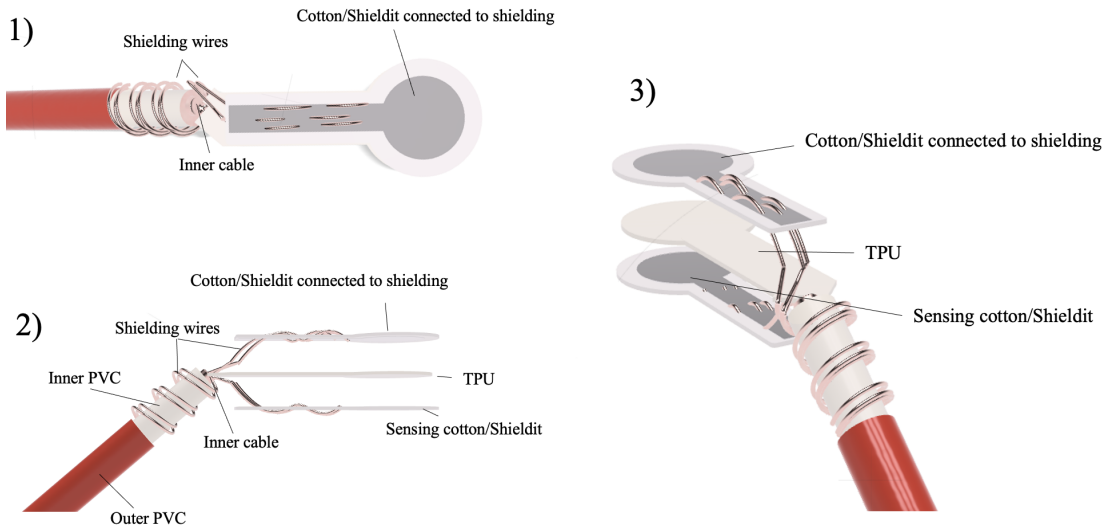


Figure 7: 1) shows the top view, 2) a lateral view and 3) an angular view of the final prototype. An upper layer of Shieldit/cotton was attached to the shielding wires by weaving the copper wires through the fabric. The bottom layer of Shieldit/cotton was attached to the inner cable in similar manner. The latter would make contact with the skin to measure biopotential signals. The two layers and their wiring were separated by an insulating layer of TPU.

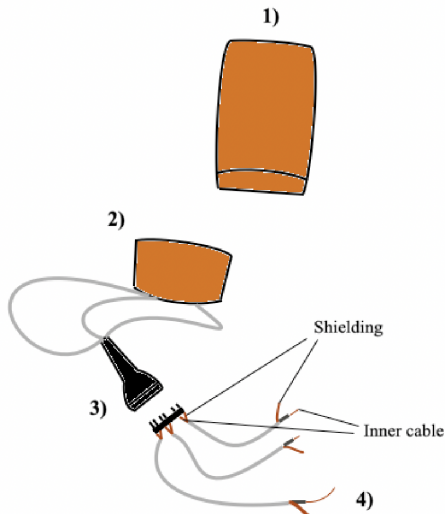


Figure 8: Customized DPHA-16 header when using shielded cables.

Electrode:

The prototype was considered to be breathable and biocompatible, since Shieldit conductive fabric had already been used in wearable sensors [18] and cotton is a common fabric in clothing and is known to be breathable. Since cadmium, mercury, chromium, PBB, PBDE, DEHP, BBP, DBP and DIBP were not used within the prototype, the prototype was considered to be RoHS compliant [24].

When considering costs, the manufacturing price of one prototype would be €0,08 (Appendix F). The costs remain below the buying price for one Ag/AgCl electrode (e.g. €0,68), so this wish was met as well. The cable was not included within this calculation, since this is not included in the price for the Ag/AgCl electrode either [14].

The diameter of the Shieldit fabric was set to 7,6 mm so that when comparing the prototype to the Ag/AgCl electrode, diameter could not be influencing signal quality. Thickness of

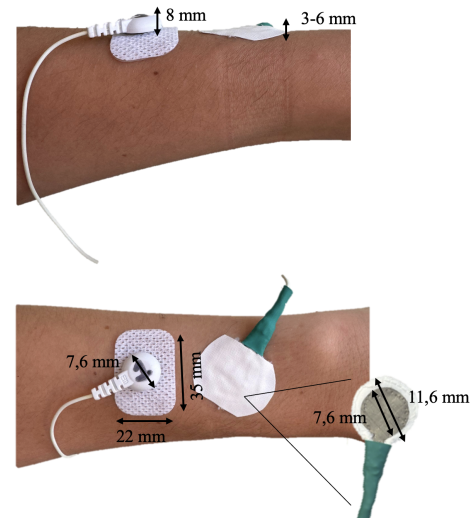


Figure 9: Dimensions of the currently used Ag/AgCl electrode (left) and the final prototype (right). The image above shows a side view and the bottom image a top view.

the electrode increased relative to the second version of the prototype due to the different layers that were incorporated. This resulted in a total thickness of the electrode of 3 mm. In the end, thickness requirements were still met.

Electrode-cable connection:

Weaving the wires through the fabric resulted in a connection between electrode and cable that was not a snap and that was flexible. As shown in Appendix D, the woven cable-connection can resist pulling forces up to 49,05 N whilst maintaining good conductivity (resistance 0,5 Ω). The two woven electrode-cable connections on top of each other resulted in a total thickness of the electrode-cable connection of 6 mm.

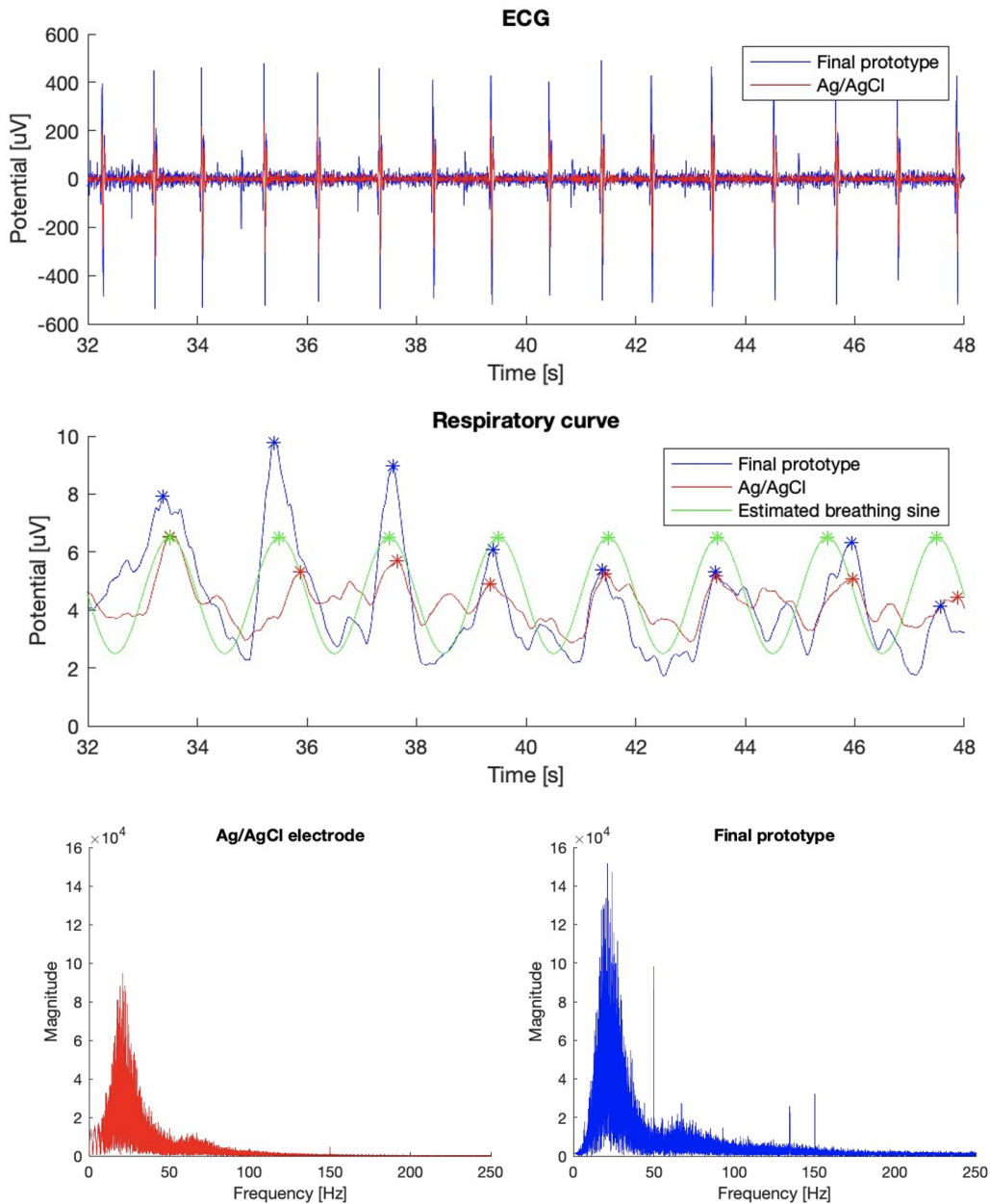


Figure 10: Above: the raw ECG measured by the Ag/AgCl electrode (red) and the final prototype (blue). In the middle: the extracted breathing curve for both electrodes. The expected breathing curve (green) was approximated by plotting a sine with a frequency of 30/min (e.g. 0,5 Hz) and arbitrary amplitude. A star denotes a detected peak in the breathing curve. The two lower graphs show the frequency plots for the Ag/AgCl electrode and the textile electrode.

Skin-electrode interface:

This band-aid had already been used in medical applications and therefore was considered to be biocompatible. In addition, no sharp edges were designed in the prototype and since the prototype was made of textile, it was considered to be soft to the skin. Since no conductive gel was used within this design, the prototype was considered to be a dry electrode. During impedance measurements with the second version of the prototype (Appendix E), the electrode did not fall of the skin spontaneously.

Cable:

The prototype used three leads, connected to the Dipha-16. Shielded cables were used and it was aimed to shield the

electrode itself as well. However, from Test C (Appendix F) followed that a 50 Hz peak still occurred in the frequency plot.

Overall:

To quantify the quality of the signal that was measured, the RMSE value relative to a known 10 Hz sine and the SD were calculated during Test C (Appendix F). The Ag/AgCl electrode had a RMSE value of 0,5998 mV and a SD of 0,0316 mV, whereas the final prototype had a RMSE value of 0,5984 mV and a SD of 0,1031 mV. The final prototype did not have a lower SD than the Ag/AgCl electrode, but the final design did perform slightly better regarding RMSE value. This meant that the measured signal by the final design was, on average, closer to the simulated 10 Hz sine. But, that there was more

variation between periods of the signal measured by the final design.

In the end, almost all design requirements were met. Since three months of iteration cycles had passed, the final design was taken into a proof of principle test.

E. Proof of principle

During the proof of principle test, dEMG measurements using the prototype and the standard Ag/AgCl electrode were performed simultaneously. After downloading the signals from Polybench, they were converted and post-processed using Matlab. The subject was asked to breathe in and out on a beat of 60/min. The curve of one breathing cycle was estimated by the green sinusoidal signal (Figure 10).

In Figure 10 the signal measured by the textile electrode (its ECG as well as its breathing curve) appeared to have a larger amplitude than the signal measured by the Ag/AgCl electrode. Subsequently, peak height of the breathing curves was calculated in a time interval between $t=10$ and $t=80$ s, where there were no outliers (Appendix F). The breathing peaks measured with the textile electrode had an average amplitude that was $0,3942 \mu\text{V}$ higher than the average amplitude measured by the Ag/AgCl electrode. However, this difference was not significant ($p = 0,0830$). Both electrodes measured 35 breaths within the mentioned time interval, which was equal to the amounts of peaks generated by the sine.

Finally, it was assessed how well both electrodes (Ag/AgCl and the prototype) met the set design requirements by creating a Harris profile, which was shown in Figure Appendix F. Shown visually by the Harris profiles, the final design met the design requirements the best with 22 points, versus 14 points for the Ag/AgCl electrode. The biggest factors that caused the difference in points, were the thickness of the electrodes, the electrode-cable connection and dryness of the electrode.

IV. DISCUSSION

In this paper, three versions of a textile electrode to measure dEMG were created. To verify whether these versions met the design requirements, various tests were performed: Test A (resistance to pulling forces and electrical resistance), Test B (impedance measurements) and Test C (signal quality). With the third and final prototype, a proof of principle test was conducted and a Harris profile was created. The final prototype showed improvements regarding electrode-cable connection, thickness, flexibility and dryness. In addition, for the final prototype a smaller RMSE value could be calculated than for the Ag/AgCl electrode. For SD the opposite held. Both the Ag/AgCl electrode as well as the final prototype could measure similar breathing frequency and no significant difference in breathing peak amplitude was found.

First, limitations to the testing methods were discussed. During Test A, direction of the pulling force influenced how much pulling force the connection could resist. The connection could be easily broken when it was pulled in a direction perpendicular to the gel surface (maximum pulling force: $0,981 \text{ N}$). However, when it was pulled in a direction

parallel to the gel surface, it was harder to disconnect the snap (maximum pulling force: $19,62 \text{ N}$). The last value was included in the report, since this was the same direction as pulling forces worked on the textile prototypes.

In Test B, the built-in impedance measurement function of the amplifier (Nicolet AT2+6, Natus Medical Incorporated, Wisconsin, USA) was used. However, impedance values depend on the frequency of the signal that is admitted to the body. In [6] the relation between impedance and frequency for a dry textile electrode is described, whereas [11] shows impedance graphs for the Ag/AgCl electrode. In both cases, impedance values decrease with increasing frequency. In case of Test B, the built-in impedance measurements of the amplifier could only be performed at 20 Hz. Therefore, impedance values are likely smaller when measured for higher frequencies.

When performing impedance measurements, an interesting phenomenon was discovered. In case of adhering the prototype to the skin with a conductive pad or conductive adhesive gel, the area of the skin that was covered with the pad or covered with gel influenced the impedance value that was measured. It was found that the larger the skin area covered with a conductive substance, the lower the impedance. Since a current always chooses the path of lowest resistance, a possible explanation could be that the current travels a longer path through the conductive substance and a shorter path through the body when a larger surface was covered with a conductive substance. When this phenomenon was discovered, the conductive substance was used only between electrode and skin and not outside the diameter of the electrode.

The lower the impedance, the better the signal quality (less noise) is expected to be. However, the results following from this research showed differently. Based on impedance measurements it was expected that the prototype combined with conductive adhesive gel would generate the best prototype signal and that the prototype combined with a band-aid would generate the noisiest signal. However, looking at the SD values (Test C) that were found, the textile electrode combined with a band-aid generated the best prototype signal. The fact that variations in impedance were not reflecting in signal quality, was due to the high input impedance of the Dipha-16 amplifier. Measured impedances were in the order of $\text{k}\Omega$ to $\text{M}\Omega$, whereas the Dipha-16 amplifier has an input impedance in the order of $\text{G}\Omega$. In addition, the gel pads are normally used to conduct electrical stimulation signals (massage equipment or ab muscle stimulation) rather than conducting a sensory signal. High precision and a signal that is not noisy is not required within these applications. Tests measuring the various adhesion methods were performed in different orders to verify the results were not caused by contamination of the copper plate or remains of a conductive substance of another measurement. In addition, the copper plate and the skin of the subject were cleaned between every measurement.

During Test C an inaccuracy in frequency of the Fluke simulator was noted. A sinusoidal signal with a frequency of 10 Hz and an amplitude of $0,5 \text{ mV}$ was simulated. However,

when processing the results, a phase shift was noted between the measured signals and the simulated sine. After using the peaks in the measured signal to calculate the frequency, a frequency of 10.1 Hz was found in all the measured signals. Therefore, the inaccuracy must have been caused by the Fluke simulator. It was not known whether there was an inaccuracy in the amplitude of the simulated signal as well.

Since sensitivity of the electrode could be influenced by the frequency that was measured, various frequencies were analyzed when setting up Test C. For the 10 Hz signals, a phase shift was calculated to align the expected sinusoidal signal with the measured signal. This sine was used to calculate RMSE values of the measured signals. However, when increasing frequency, this piece of Matlab code (Appendix C) would not always return the correct frequency for the simulated sine and no relationship between frequency and RMSE or SD could be established. Therefore, for higher frequencies, the phase shift was obtained by inspecting the graphs. When designing Test C, it was chosen to analyze the signals only at 10 Hz, since it was the easiest to fit a sinusoidal signal.

When analyzing the frequency plots of Test C and the proof of principle tests, a 50 Hz (and 150 Hz) peak was still picked up by both electrodes. Since the peak was picked up by both electrodes, the requirement stating a 50 Hz peak must be completely absent from the frequency plot might have been too rigorous. However, the final prototype picked up the 50 Hz frequency with higher amplitude than the Ag/AgCl electrode in both tests. It is strange that these peaks occur in the frequency plot, since lead 1 and lead 2 are subtracted from each other and should both measure the same 50 Hz signal, therefore eliminating the peak when subtracting. Various reasons why this 50 Hz peak still occurs in the frequency plots were thought of.

First, the object/subject that is measured on could pick up the 50 Hz from the mains electricity and serves as an antenna itself. Test C was measured on a copper plate that could pick up some 50 Hz from the environment and during the proof of principle test the body serves as an antenna. Interestingly, the magnitude of the 50 Hz peak in the proof of principle test is for both electrodes much larger than in Test C. In Test C, the 50 Hz peak measured by the final prototype had a magnitude of 567,7 and the Ag/AgCl electrode measured the peak with a magnitude of 126,4. During the proof of principle test, the magnitude of the peak measured by the textile electrode was 98272,7 and the magnitude of the peak measured by the Ag/AgCl electrode was 14489,2. This could be due to the fact that the body is much larger than the copper plate.

In addition, the Ag/AgCl electrode itself is not shielded, only the cable. Therefore, it will pick up some 50 Hz at the electrode as well. In the design of the final prototype, it was tried to make the electrode itself shielded, but shielding works only by creating a Faraday's cage within the spiral shield. When connecting the Shieldex conductive fabric to the spiral shield, the conductive fabric was only present on the top of the measuring surface of the electrode and therefore, did not create a cage.

Another potential cause of the 50 Hz peak is the skin-

electrode interface. Since, the magnitude of the 50 Hz peak differs between each adhesion method (Appendix (E)), it was expected to be of influence. In addition, subtraction of lead 1 and lead 2 could leave a 50 Hz peak if one of the electrodes was not attached to the body properly or one of the leads short-circuits. Since the prototype electrodes are made by hand, there could have been inaccuracies between them, influencing the subtraction.

Lastly, the pin headers used to connect the cables to the Dipha-16 were not shielded and were exposed to electromagnetic interference from the environment as well. However, this is the same for both the Ag/AgCl electrode and the final prototype.

The peak occurs in both Test C as well as the proof of principle test, indicating that picking up the 50 Hz peak does not happen due to testing set-up. Both the Ag/AgCl electrode and the final prototype were connected to the Dipha-16 using exposed pin-headers. In the end, it is believed that the 50 Hz peak most likely finds its cause in the design of the prototype or inequalities in the adhesion from electrode to skin.

When analyzing the graph following from the proof of principle test, it was noticed that the amplitude of the signal measured by the prototype was larger than the amplitude measured by the Ag/AgCl electrode, however not significant. dEMG measurements with both electrodes were performed simultaneously, so the electrodes were placed next to each other on the thorax. The location of the reference electrode on the sternum was not expected to influence the results in that manner. The placement of the active electrodes however, could influence the amplitude of the signal. The potential difference between the reference electrode and each active electrode was measured, resulting in a biopotential measurement for lead 1 and a measurement for lead 2. By subtracting these leads the prototype and the Ag/AgCl electrode measure in the same transversal plane in the thorax. On the sides, the diaphragm is almost vertically shaped, so it was not expected that the prototype has measured a lot more diaphragm muscle than the Ag/AgCl electrode. In addition, [17] showed that placing electrodes in the same transversal plane should not influence amplitude significantly. However, changing the position of the electrode in vertical direction is of influence on the signal amplitude. Another argument why a larger amplitude is likely not due to placement of electrodes, is that these findings were consistent with findings from Test B, where the amplitude measured by the textile electrode was larger than the amplitude measured by the Ag/AgCl electrode as well. Therefore, the influence of the electrode placement during the proof of principle test is expected to be minimal. The proof of principle test was performed on a healthy adult female. Since adults have larger bodies and thicker skin than neonates, it is unsure whether results found in healthy adults can be extended to neonates at the NICU. In the end, the results of the proof of principle test look very promising. However, further improvements and further tests on healthy adults as well as on patients need to be performed.

Current ways of measuring non-invasive dEMG in preterm infants are either using standard Ag/AgCl electrodes or a

sensor belt such as the Bambi belt. From the introduction, the downsides of both these methods should be clear.

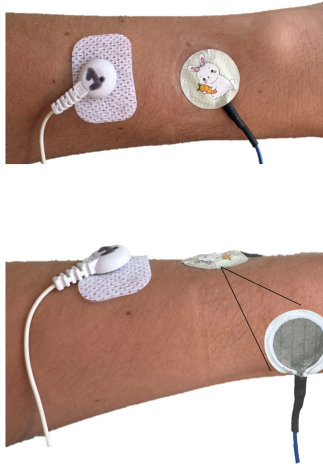


Figure 11: What a pediatric textile electrode with a shielded cable could look like. If a band-aid is used as a basis, the Shieldit Super conductive fabric could be ironed on the cotton cushion of the band-aid. The shielded cable could be woven into the Shieldit-Super fabric and a child-friendly picture can be added to the back of the band-aid.

To still be able to place the results of this paper into literature context, first current methods to measure ECG in preterm infants were assessed, followed by methods to measure EMG using textile electrodes.

In [7] a textile jacket using textile electrodes made of Shieldex conductive fabric was described. This jacket was used to measure ECG in neonates. In [8] respiration in neonates is measured using a PDMS-Graphene strain sensor (measuring thorax excursions) and ECG is measured using textile (Shieldex) electrodes. These sensors are integrated into a vest that provides cardiorespiratory information. However, measuring chest excursions using a strain sensor was not preferred in this paper as a method to measure respiratory state, since the thorax can expand without actual airflow (for instance in case of an obstruction). In addition, a vest was not preferred, since this would comprise a large area of the babies chest. Other options to measure ECG in neonates are integrating textile electrodes or capacitive arrays into a mattress. However, this is considered to be unfit to measure dEMG due to the placement of the electrodes relative to the diaphragm. Lastly, neonatal or pediatric ECG electrodes exist. These are pre-wired Ag/AgCl electrodes that are created for neonates specifically. However, the wires that are used in these electrodes are unshielded. Even if a shielded cable was connected to these pediatric electrodes, this paper showed that a dry textile electrode could have various advantages over an Ag/AgCl electrode.

Regarding EMG measurements in general using textile electrodes, it was found that most textile electrodes measuring EMG find their application in wearables. Most of these wearables are used in sports or rehabilitation related applications. A challenge for the textile prototype in this paper was adhesion to the skin. It was undesired to design something that would comprise a large area of the babies chest, so a wearable shirt or onesie was not an option.

Summarizing, currently used non-invasive methods to measure dEMG in preterm infants have their downsides. Some methods that are currently used to measure ECG in preterm infants cannot be used to measure dEMG as well. Textile electrodes are already used in some ECG applications in neonates, which can be of inspiration. In this paper, using textile electrodes in combination with a shielded cable to measure dEMG is something that has been proven to work and to generate a good quality signal.

For future recommendations, it would be advised to do more research to find the source of the 50 Hz peak in the frequency plot by testing various substrate materials for the electrode (PVC or a polymer for example), testing other adhesion methods and exploring various testing set-ups. Since shielding did not work, could be removed in post-processing and only increased thickness of the electrode and the electrode-cable connection, another possibility is to solely shield the cable and not the electrode itself in a future design. 11 shows what a product using a shielded cable, a textile sensing surface and a band-aid as fixation could look like.

A current advantage to the Ag/AgCl electrode is that the snap connection can turn. So, if the cable is handled with care, the cable can be turned around an axis perpendicular to the skin. Therefore, an idea for a new iteration cycle would be to design a flexible cable connection where the cable is able to pivot.

One of the design wishes for this prototype was a small diameter. To correct for influence of electrode diameter, the diameter of the prototype was kept equal to the diameter of the sensing surface of the Ag/AgCl electrode. However, in the future more research could be done to find the minimum diameter the electrode could have whilst still obtaining good signal quality. If possible, electrodes with various diameters could be fabricated and customized to the patient.

As mentioned, an important condition was to comprise an area of the thorax as small as possible. The reason only a small area of the neonate's thorax could be comprised, was the presence of ECG electrodes, visibility of the thorax during a clinical assessment of the patient and uncomfortable sight for the parents. Therefore, integrating the dEMG electrodes into a patch, a vest, a romper, etcetera was not an option. Integrating all desired biopotential sensors (ECG as well as dEMG) in one wireless, friendly-looking, patch, could be a way to overcome both obstacles in the future.

Another thing that was ruled out by the desire for a small surface, was the wirelessness of the separate electrodes. In order to measure a biopotential signal, a ground electrode and an active electrode are needed. The biopotential signal is measured by measuring the difference between the two. Therefore, these electrodes need to be connected in some manner. If the electrode should become wireless, a small connection between active electrode and the reference electrode has to remain. Another challenge when designing a wireless electrode is the battery. The battery adds to the thickness and detracts from the flexibility of the electrode. A wireless electrodes is very promising, however a few design hurdles still need to be overcome.

V. CONCLUSION

The goal of the paper was to develop a prototype electrode that could measure dEMG in preterm infants. The prototype had to be dry, thinner and more flexible than the currently used Ag/AgCl electrode, whilst maintaining the same signal quality.

Ultimately, not all design requirements were met. The frequency plot of the final prototype still showed a peak at 50 Hz, indicating insufficient shielding from electromagnetic interference. However, the final prototype was indeed dry, thinner, and more flexible than the Ag/AgCl electrode. The RMSE value for the prototype was smaller than that for the Ag/AgCl electrode, although the opposite was true for the SD. In addition, it was proven that the final prototype could be used to measure dEMG in healthy adults. The breathing frequency measured by the final prototype was equal to the breathing frequency measured by the Ag/AgCl electrode, whilst showing no significant difference in amplitude.

In conclusion, it has been proven that a textile prototype is well capable of measuring dEMG in humans and offers several improvements over the Ag/AgCl electrode. However, further research is necessary to extend these findings to preterm infants and further improve the prototype.

July 23, 2024

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APPENDIX A
NON-WMO STATEMENT

Title of the Study: Creating a Thin, Flexible Biopotential Electrode to Measure dEMG in Preterm Infants

Researchers: M. Stulen

Date: July 23rd, 2024

After a thorough review of the study protocol, I have determined that the proposed study does not fall under the scope of the Dutch Medical Research Involving Human Subjects Act (Wet Medisch-wetenschappelijk Onderzoek met mensen, WMO), as defined by the law and the guidelines of the Central Committee on Research Involving Human Subjects (Centrale Commissie Mensgebonden Onderzoek, CCMO). The study involves self-measurements conducted by the author and does not include any interventions or procedures involving other human subjects.

Reasons for Non-WMO Status:

- 1) The study involves only self-measurements by the author, with no involvement of other human subjects.
- 2) No physical or psychological interventions are performed on other participants.
- 3) The self-measurements do not affect the medical treatment of the author or any other individuals.
- 4) The collected data are solely intended for personal and academic purposes and are not shared with or traceable to other individuals.

Based on the above considerations, I conclude that this study is not subject to WMO regulations and therefore does not require review by a Medical Ethics Review Committee (METC).

M. Stulen

Student Mechanical Engineering (track: BioMechanical Design) at the TU Delft

APPENDIX B TESTING METHODS

A. Test A

The resistance to pulling forces was defined as the maximum pulling force the electrode-cable connection could handle without breaking. Therefore, this resistance was expressed in Newtons. Pulling force was measured by attaching a weight to the cable of the prototype. Then, the electrode was held in the air by its sensing surface. First, the smallest weight was attached to the pin header, after which the weight was increased until the electrode or electrode-cable connection would break or maximum weight was reached. The weight that was used when the electrode or electrode-cable connection would break, was recorded and multiplied by $9,81 \text{ m/s}^2$ to calculate a force. The weights that were used were 10g, 20g, 100g, 1,25 kg, 2 kg, 2,5 kg, 3 kg and 5 kg (respectively 0,0981 N; 0,1962 N; 0,981 N; 12,26 N; 19,62 N; 24,53 N; 29,43 N; 49,05 N).

The electrical resistance of a cable-electrode connection was measured using a digital multimeter (Fluke 15B+, Fluke, Washington, USA). One end of the multimeter was held against the sensing surface of the electrode and the other end was held against the end of the cable. The electrical resistance given by the multimeter was recorded.

Finally, a table was created showing breaking force and electrical resistance before the test was performed. Of the cable-electrode connections that did not break, the electrical resistance after the test was noted as well.

B. Test B

To measure impedance of the skin-electrode interface, a brain monitor (Xltek, Natus Medical Incorporated, Wisconsin, USA) in combination with an amplifier (Nicolet AT2+6, Natus Medical Incorporated, Wisconsin, USA) and its built-in impedance measurement system was used. Normally, measurements using these devices are performed to assess whether EEG silver cup electrodes make good contact with the patient's skin.

In this paper, one of the silver cup electrodes was replaced by the electrode (prototype or standard Ag/AgCl) of interest, as shown in Figure 12. The software program would return a value for impedance in $k\Omega$ (Figure 12), which was reported. The electrode of interest (prototype or Ag/AgCl) was attached to the skin using various adhesion methods. To attach the prototype to the skin a band aid, conductive adhesive gel [20] or conductive adhesive pads [1] could be used. The electrodes were placed on the lower arm of a 25 year old healthy female volunteer, as shown in Figure 12, and measurements were repeated multiple times.

Between every measurement repetition the electrode of interest was removed, the skin was cleaned using ethanol and the electrode was placed back on the same location using a different adhesion method. This cycle was repeated until every adhesion method was tested 5 times. Then, the skin was cleaned and scrubbed using a scrub gel (NUPREP [19], Weaver and Company, Colorado, USA) and a cotton bud.

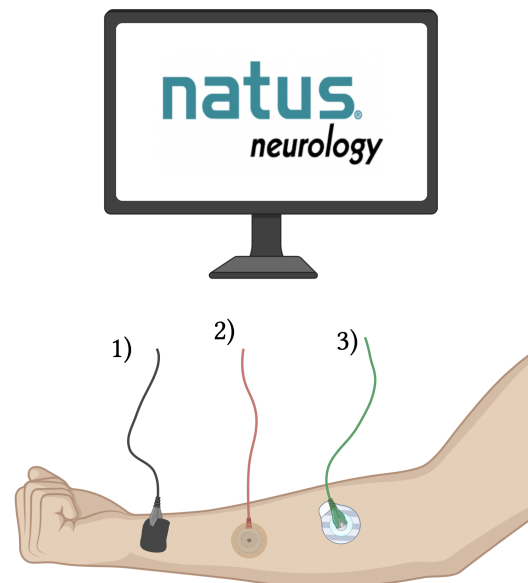


Figure 12: A schematic drawing of impedance measurements, where 1) is the reference electrode, 2) a silver cup electrode and 3) the electrode of interest.

Excessive scrub was removed using a dry tissue and the 5 measurements per electrode in combination with a specific adhesion method were repeated.

This resulted in 10 measurements per electrode and a certain adhesion method: 5 with skin preparation and 5 without skin preparation.

C. Test C

To quantify the quality of the measured signal, a simulator (FLUKE ESA620 230V, Fluke, Washington, USA) was used to simulate a 10 Hz sine with an amplitude of 0,5 mV. A copper printed circuit board (PCB) was connected to the simulator. On this PCB, electrodes could be attached so that the simulated signal could be measured (Figure 13).

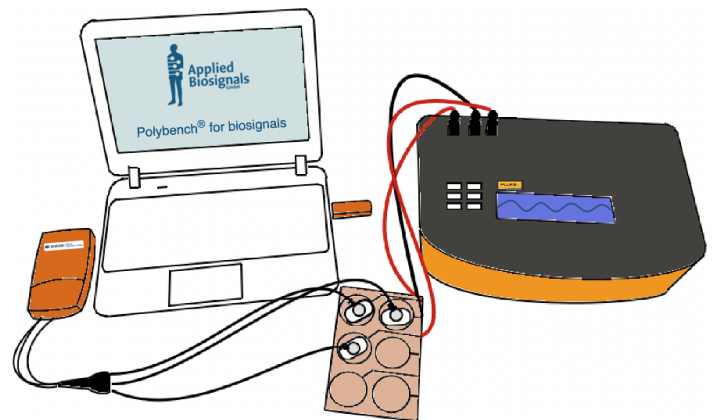


Figure 13: Testing set-up to measure a simulated sine 10 Hz. The copper PCB in the middle was connected to the simulator. The electrodes could be placed on the copper PCB and were connected to the DIPA-16 amplifier (orange). In the end, Polybench was used to visualize the signal.

These measurements were performed with the Ag/AgCl electrode as well as the prototype. The prototype was attached

to the copper PCB using various adhesion methods: a band-aid, conductive pads or conductive adhesive gel. Measurements with each electrode configuration were performed for 1 minute. Then, the measured periods of the signal were separated and layed over each other using a Matlab script (Appendix C). Since the input was a known 10 Hz sine, the root mean square error (RMSE) value and the standard deviation (SD) of all these overlayed periods relative to the known sine was calculated and the average value was taken to come to one RMSE value and one SD.

The RMSE value is used to measure the distance between a measured array (F_i , where i a specific time instance in one period) and a true array (10 Hz sinusoidal signal A_i , where i is the same time instance as in F_i). Therefore, the RMSE value shows how well a certain measurement approaches the true signal. The RMSE value for one period was calculated using the following formula [25]:

$$E = \sqrt{\frac{1}{n} \sum_{i=1}^n |A_i - F_i|^2}$$

After calculating the RMSE value for every period, the average over all periods was taken to provide one RMSE value for each electrode configuration.

The SD shows the spread of the data A_i around its mean μ and was calculated for each period as well, using the following formula:

$$S = \sqrt{\frac{1}{N-1} \sum_{i=1}^N |A_i - \mu|^2}$$

and μ :

$$\mu = \frac{1}{N} \sum_{i=1}^N A_i$$

After calculating the SD for every period, the average was taken over all periods to provide one value for the SD per electrode configuration.

The cable was considered to be shielded when a 50 Hz peak from the mains electricity would not occur in a frequency plot, which was created by the Matlab (Appendix C) script as well.

APPENDIX C

MATLAB CODE TEST C

%% Graphs

% Sequence: ECG, textile, textile + pad, textile + gel @ 10 Hz

%% ECG

clear all

close all

clc

run Poly5_Read.m

%%%%%%%%%%%%%% Time

t_end=length(PolyCurves)/SampleFreq;

tA=linspace(0,t_end,SampleFreq*t_end);

%%%%%%%%%%%%%% Raw ECG data

Delta_12=PolyCurves(:,1)-PolyCurves(:,2);

EMG1=PolyCurves(:,1);

EMG2=PolyCurves(:,2);

Delta_12_filt_highA=Delta_12./1000; % microV to miliV, difference between lead 1 and lead 2

%%%%%%%%%%%%%% Expected simulated sine

freq=10;

A=0.5; % mV, amplitude

dt=0; % s, can be adjusted

S=A*sin(freq*2*pi*(tA-dt));

%%%%%%%%%%%%%% Finding the equilibrium of the signal

min1=min(Delta_12_filt_highA(5000:10000));

max1=max(Delta_12_filt_highA(5000:10000));

dA=(min1+max1)/2; % calculates the equilibrium stance

Delta_12_filt_highA=Delta_12_filt_highA-dA;

%%%%%%%%%%%%%% Find local peaks

M1=movmean(Delta_12_filt_highA, 15);

M2=bandpass(M1, [freq-1 freq+1], SampleFreq); % removing faster noisy frequencies

[pks1, locs1]=findpeaks(M2, 'MinPeakDistance', 0.8*(1/freq)); % finding the peaks of the mean

[pks3, locs3]=findpeaks(Delta_12_filt_highA, 'MinPeakDistance', 0.8*(SampleFreq/freq)); % finding the peaks of the original signal

%%%%%%%%%%%%%% Finding period of the signal

for i=1:(length(locs3)-1)

Diff(i)=locs3(i+1)-locs3(i); % determines the distance between peaks

end

Nmeas=mean(Diff); % mean distance between the peaks in # samples

Tmeas=Nmeas*(1/SampleFreq);

fmeas=1/Tmeas; % returns the measured frequency of the signal

%%%%%%%%%%%%%% Expected simulated sine

A=0.5; % mV, amplitude

dt=0; % s, nog aanpassen

S=A*sin(fmeas*2*pi*(tA-dt)); % Fitting a sine to the signal.

%%%%%%%%%%%%%% Finding the phase shift between expected sine and

%%%%%%%%%%%%%% measured signal

[pks2, locs2]=findpeaks(S);

```

B=2*pi*fmeas;

if locs2(1)>=locs1(1)
    dt=tA(locs2(1))-tA(locs1(1));
    SA=A*sin(B*(tA+dt));
elseif locs2(1)<=locs1(1);
    dt=tA(locs1(1))-tA(locs2(1));
    SA=A*sin(B*(tA-dt));
else
    SA=0;
end

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Creating the overlay plot and RMSE value
Nmeas1=floor(Nmeas); % floored average # samples between peaks
CROSS=zeros(100,100); % With CROSS the columns corresponds to the single periodes:
    from peak n till peak n+1, the rows corresponds to the measurement values. This
    matrix is set before hand because every period has different length.
SAT=SA'; % transposing the sine to calculate RMSE value later

figure()
hold on
for i=1:(length(pks3)+1-floor(Nmeas/4)) %1/4 from the period is subtracted from the
    peaks, so the value at equilibrium is found
    A(i)=(locs3(i+1)-floor(Nmeas/4))-(locs3(i)-floor(Nmeas/4))+1; % delivers the
    length of the i'th period
    CROSS(1:A(i), i)=Delta_12_filt_highA((locs3(i)-floor(Nmeas/4)):(locs3(i+1)-floor(
    Nmeas/4))); % fills the cross matrix.
    RMSE(i,1)=rmse(CROSS(1:A(i),i), SAT(1:A(i),1)); % calculates the RMSE value for
    every column with a different length.
    plot(tA(1,1:A(i)), CROSS(1:A(i),i))
end

RMSE_tot_A=mean(RMSE);

for i=1:100
    MEAN(i)=mean(nonzeros(CROSS(i,:))); % calculating the mean of every row, without
    taking into account the zeros that are in their because not all columns have
    equal length
    STD(i)=std(nonzeros(CROSS(i,:)));
end

M3=movmean(MEAN, 17);
%A=max(M3)?
Amp=0.5;
S2=(Amp)*sin(fmeas*2*pi*(tA)); % the sine that will be in the overlay plot
plot(tA(1:100), MEAN, 'r', 'LineWidth',3) % mean of all the measured periods, -dA to
    correct for the offset.
plot(tA(1:100), S2(1:100), 'g', 'LineWidth',3) % expected sine
title('Ag/AgCl electrode @ 10 Hz')
xlabel('Time [s]')
ylabel('Potential [mV]')
xlim([0 0.1])
ylim([-3 3])

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Calculating the 95% confidence interval
Upper=MEAN+STD.*1.98;
Lower=MEAN-STD.*1.98;

```

```

plot(tA(1:100), Upper, 'b', 'LineWidth',2)
plot(tA(1:100), Lower, 'b', 'LineWidth',2)

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Frequency plot of the signal
fftSignal = fft(Delta_12_filt_highA);
fftSignal = fftshift(fftSignal);

dF = SampleFreq/length(PolyCurves(:,1));
f = -SampleFreq/2:dF:SampleFreq/2-dF;

figure()
hold on
plot(f, abs(fftSignal), 'b')
title('Ag/AgCl electrode @ 10 Hz')
xlabel('Frequency [Hz]')
ylabel('Magnitude')
ylim([0 5000])

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Plotting the total signal and all the relevant points
figure()
hold on
plot(tA, Delta_12_filt_highA, 'b')
plot(tA(locs1), pks1, 'k*')
plot(tA(locs2), pks2, 'm*')
plot(tA, SA, 'r')
plot(tA(locs3), pks3, 'y*')
plot(tA, M2, 'g')
ylim([-1 1])
legend('Measured signal', 'Peaks of bandpassed moving mean', 'Peaks of the expected
      sine (without phase shift)', 'Expected sine (phase shifted)', 'Peaks of the
      measured signal', 'Bandpassed moving mean of the measured signal')

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Plotting in the subplot
figure(13)
hold on
subplot(4,1,1)
hold on
plot(tA-0.05, SA, 'r')
plot(tA-0.05, Delta_12_filt_highA, 'b')
xlim([0.1 2.1])
ylim([-2 2])
xlabel('Time [s]')
ylabel('Potential [mV]')
title('Ag/AgCl electrode')
legend('Simulated signal', 'Measured signal')

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Returning overall STD
STD(isnan(STD))=0;
STD_all_A=mean(nonzeros(STD));

%% Textile electrode
clear all
clc
run Poly5_Read.m

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Time
t_end=length(PolyCurves)/SampleFreq;
tB=linspace(0,t_end,SampleFreq*t_end);

```



```

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Raw ECG data
Delta_12=PolyCurves(:,1)-PolyCurves(:,2);
EMG1=PolyCurves(:,1);
EMG2=PolyCurves(:,2);
Delta_12_filt_highB=Delta_12./1000; % microV to miliV, difference between lead 1 and
    lead 2

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Expected simulated sine
freq=10;
A=0.5; % mV, amplitude
dt=0; % s, nog aanpassen
S=A*sin(freq*2*pi*(tB-dt));

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Finding the equilibrium of the signal
min1=min(Delta_12_filt_highB(5000:10000));
max1=max(Delta_12_filt_highB(5000:10000));
dA=(min1+max1)/2; % calculates the equilibrium stance
Delta_12_filt_highB=Delta_12_filt_highB-dA;

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Find local peaks
M1=movmean(Delta_12_filt_highB, 15);
M2=bandpass(M1, [freq-1 freq+1], SampleFreq); % removing faster noisy frequencies
[pks1, locs1]=findpeaks(M2, 'MinPeakDistance', 0.8*(1/freq)); % finding the peaks of
    the mean
[pks3, locs3]=findpeaks(Delta_12_filt_highB, 'MinPeakDistance', 0.8*(SampleFreq/freq)
    ); % finding the peaks of the original signal

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Finding period of the signal
for i=1:(length(locs3)-1)
    Diff(i)=locs3(i+1)-locs3(i); % determines the distance between peaks
end
Nmeas=mean(Diff); % mean distance between the peaks in # samples
Tmeas=Nmeas*(1/SampleFreq);
fmeas=1/Tmeas; % returns the measured frequency of the signal

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Expected simulated sine
A=0.5; % mV, amplitude
dt=0; % s, nog aanpassen
S=A*sin(fmeas*2*pi*(tB-dt)); % Fitting a sine to the signal.

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Finding the phase shift between expected sine and
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% measured signal
[pks2, locs2]=findpeaks(S);
B=2*pi*fmeas;

if locs2(1)>=locs1(1)
    dt=tB(locs2(1))-tB(locs1(1));
    SB=A*sin(B*(tB+dt));
elseif locs2(1)<=locs1(1);
    dt=tB(locs1(1))-tB(locs2(1));
    SB=A*sin(B*(tB-dt));
else
    SB=0;
end

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Creating the overlay plot and RMSE value
Nmeas1=floor(Nmeas); % floored average # samples between peaks

```

```

CROSS=zeros(100,100); % With CROSS the columns corresponds to the single periodes:
    from peak n till peak n+1, the rows corresponds to the measurement values. This
    matrix is set before hand because every period has different length.
SBT=SB'; % transposing the sine to calculate RMSE value later

figure()
hold on
for i=1:1:(length(pks3)+1-floor(Nmeas/4)) %1/4 from the period is subtracted from the
    peaks, so the value at equilibrium is found
    A(i)=(locs3(i+2)-floor(Nmeas/4))-(locs3(i+1)-floor(Nmeas/4))+1; % delivers the
    length of the i'th period
    CROSS(1:A(i), i)=Delta_12_filt_highB((locs3(i+1)-floor(Nmeas/4)):(locs3(i+2)-
    floor(Nmeas/4))); % fills the cross matrix.
    RMSE(i,1)=rmse(CROSS(1:A(i),i), SBT(1:A(i),1)); % calculates the RMSE value for
    every column with a different length.
    plot(tB(1,1:A(i)), CROSS(1:A(i),i))
end

RMSE_tot_B=mean(RMSE);

for i=1:100
    MEAN(i)=mean(nonzeros(CROSS(i,:))); % calculating the mean of every row, without
    taking into account the zeros that are in their because not all columns have
    equal length
    STD(i)=std(nonzeros(CROSS(i,:)));
end

M3=movmean(MEAN, 17);
%A=max(M3)?
Amp=0.5;
S2=(Amp)*sin(fmeas*2*pi*(tB)); % the sine that will be in the overlay plot
plot(tB(1:100), MEAN, 'r', 'LineWidth',3) % mean of all the measured periods, -dA to
    correct for the offset.
plot(tB(1:100), S2(1:100), 'g', 'LineWidth',3) % expected sine
title('Textile electrode + band-aid @ 10 Hz')
xlabel('Time [s]')
ylabel('Potential [mV]')
xlim([0 0.1])
ylim([-3 3])

%%%%%%%%%%%% Calculating the 95% confidence interval
Upper=MEAN+STD.*1.98;
Lower=MEAN-STD.*1.98;

plot(tB(1:100), Upper, 'b', 'LineWidth',2)
plot(tB(1:100), Lower, 'b', 'LineWidth',2)

%%%%%%%%%%%% Frequency plot of the signal
fftSignal = fft(Delta_12_filt_highB);
fftSignal = fftshift(fftSignal);

dF = SampleFreq/length(PolyCurves(:,1));
f = -SampleFreq/2:dF:SampleFreq/2-dF;

figure()
hold on
plot(f, abs(fftSignal), 'b')
title('Textile electrode + band-aid @ 10 Hz')

```

```

xlabel('Frequency [Hz]')
ylabel('Magnitude')
ylim([0 5000])

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Plotting the total signal and all the relevant points
figure()
hold on
plot(tB, Delta_12_filt_highB, 'b')
plot(tB(locs1), pks1, 'k*')
plot(tB(locs2), pks2, 'm*')
plot(tB, SB, 'r')
plot(tB(locs3), pks3, 'y*')
plot(tB, M2, 'g')
ylim([-1 1])
legend('Measured signal', 'Peaks of bandpassed moving mean', 'Peaks of the expected
      sine (without phase shift)', 'Expected sine (phase shifted)', 'Peaks of the
      measured signal', 'Bandpassed moving mean of the measured signal')

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Plotting in the subplot
figure(13)
hold on
subplot(4,1,2)
hold on
plot(tB, SB, 'r')
plot(tB, Delta_12_filt_highB, 'b')
xlim([0.1 2.1])
ylim([-2 2])
xlabel('Time [s]')
ylabel('Potential [mV]')
title('Textile electrode + band-aid')
legend('Simulated signal', 'Measured signal')

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Returning overall STD
STD(isnan(STD))=0;
STD_all_B=mean(nonzeros(STD));

%% Textile electrode + pad
clear all
clc
run Poly5_Read.m

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Time
t_end=length(PolyCurves)/SampleFreq;
tC=linspace(0,t_end,SampleFreq*t_end);

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Raw ECG data
Delta_12=PolyCurves(:,1)-PolyCurves(:,2);
EMG1=PolyCurves(:,1);
EMG2=PolyCurves(:,2);
Delta_12_filt_highC=Delta_12./1000; % microV to miliV, difference between lead 1 and
      lead 2

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Expected simulated sine
freq=10;
A=0.5; % mV, amplitude
dt=0; % s, nog aanpassen
S=A*sin(freq*2*pi*(tC-dt));

```

```

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Finding the equilibrium of the signal
min1=min(Delta_12_filt_highC(5000:10000));
max1=max(Delta_12_filt_highC(5000:10000));
dA=(min1+max1)/2; % calculates the equilibrium stance
Delta_12_filt_highC=Delta_12_filt_highC-dA;

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Find local peaks
M1=movmean(Delta_12_filt_highC, 15);
M2=bandpass(M1, [freq-1 freq+1], SampleFreq); % removing faster noisy frequencies
[pks1, locs1]=findpeaks(M2, 'MinPeakDistance', 0.8*(1/freq)); % finding the peaks of
the mean
[pks3, locs3]=findpeaks(Delta_12_filt_highC, 'MinPeakDistance', 0.8*(SampleFreq/freq)
); % finding the peaks of the original signal

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Finding period of the signal
for i=1:(length(locs3)-1)
    Diff(i)=locs3(i+1)-locs3(i); % determines the distance between peaks
end
Nmeas=mean(Diff); % mean distance between the peaks in # samples
Tmeas=Nmeas*(1/SampleFreq);
fmeas=1/Tmeas; % returns the measured frequency of the signal

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Expected simulated sine
A=0.5; % mV, amplitude
dt=0; % s, nog aanpassen
S=A*sin(fmeas*2*pi*(tC-dt)); % Fitting a sine to the signal.

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Finding the phase shift between expected sine and
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% measured signal
[pks2, locs2]=findpeaks(S);
B=2*pi*fmeas;

if locs2(1)>=locs1(1)
    dt=tC(locs2(1))-tC(locs1(1));
    SC=A*sin(B*(tC+dt));
elseif locs2(1)<=locs1(1);
    dt=tC(locs1(1))-tC(locs2(1));
    SC=A*sin(B*(tC-dt));
else
    SC=0;
end

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Creating the overlay plot and RMSE value
Nmeas1=floor(Nmeas); % floored average # samples between peaks
CROSS=zeros(100,100); % With CROSS the columns corresponds to the single periodes:
    from peak n till peak n+1, the rows corresponds to the measurement values. This
    matrix is set before hand because every period has different length.
SCT=SC'; % transposing the sine to calculate RMSE value later

figure()
hold on
for i=1:1:(length(pks3)+1-floor(Nmeas/4)) %1/4 from the period is subtracted from the
    peaks, so the value at equilibrium is found
    A(i)=(locs3(i+2)-floor(Nmeas/4))-(locs3(i+1)-floor(Nmeas/4))+1; % delivers the
    length of the i'th period
    CROSS(1:A(i), i)=Delta_12_filt_highC((locs3(i+1)-floor(Nmeas/4)):(locs3(i+2)-
    floor(Nmeas/4))); % fills the cross matrix.

```

```

    RMSE(i,1)=rmse(CROSS(1:A(i),i), SCT(1:A(i),1)); % calculates the RMSE value for
    every column with a different length.
    plot(tC(1,1:A(i)), CROSS(1:A(i),i))
end

RMSE_tot_C=mean(RMSE);

for i=1:100
    MEAN(i)=mean(nonzeros(CROSS(i,:))); % calculating the mean of every row, without
    taking into account the zeros that are in their because not all columns have
    equal length
    STD(i)=std(nonzeros(CROSS(i,:)));
end

M3=movmean(MEAN, 17);
% A=max(M3)?
Amp=0.5;
S2=Amp*sin(fmeas*2*pi*(tC)); % the sine that will be in the overlay plot
plot(tC(1:100), MEAN, 'r', 'LineWidth',3) % mean of all the measured periods, -dA to
    correct for the offset.
plot(tC(1:100), S2(1:100), 'g', 'LineWidth',3) % expected sine
title('Textile electrode + conductive pad @ 10 Hz')
xlabel('Time [s]')
ylabel('Potential [mV]')
xlim([0 0.1])
ylim([-3 3])

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Calculating the 95% confidence interval
Upper=MEAN+STD.*1.98;
Lower=MEAN-STD.*1.98;

plot(tC(1:100), Upper, 'b', 'LineWidth',2)
plot(tC(1:100), Lower, 'b', 'LineWidth',2)

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Frequency plot of the signal
fftSignal = fft(Delta_12_filt_highC);
fftSignal = fftshift(fftSignal);

dF = SampleFreq/length(PolyCurves(:,1));
f = -SampleFreq/2:dF:SampleFreq/2-dF;

figure()
hold on
plot(f, abs(fftSignal), 'b')
title('Textile electrode + conductive pad @ 10 Hz')
xlabel('Frequency [Hz]')
ylabel('Magnitude')
ylim([0 5000])

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Plotting the total signal and all the relevant points
figure()
hold on
plot(tC, Delta_12_filt_highC, 'b')
plot(tC(locs1), pks1, 'k*')
plot(tC(locs2), pks2, 'm*')
plot(tC, SC, 'r')
plot(tC(locs3), pks3, 'y*')
plot(tC, M2, 'g')

```

```

legend('Measured signal', 'Peaks of bandpassed moving mean', 'Peaks of the expected
      sine (without phase shift)', 'Expected sine (phase shifted)', 'Peaks of the
      measured signal', 'Bandpassed moving mean of the measured signal')
ylim([-2 2])

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Plotting in the subplot
figure(13)
hold on
subplot(4,1,3)
hold on
plot(tC, SC, 'r')
plot(tC, Delta_12_filt_highC, 'b')
xlim([0.1 2.1])
ylim([-2 2])
xlabel('Time [s]')
ylabel('Potential [mV]')
title('Textile electrode + pad')
legend('Simulated signal', 'Measured signal')

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Returning overall STD
STD(isnan(STD))=0;
STD_all_C=mean(nonzeros(STD));

%% Textile + gel
clear all
clc
run Poly5_Read.m

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Time
t_end=length(PolyCurves)/SampleFreq;
tD=linspace(0,t_end,SampleFreq*t_end);

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Raw ECG data
Delta_12=PolyCurves(:,1)-PolyCurves(:,2);
EMG1=PolyCurves(:,1);
EMG2=PolyCurves(:,2);
Delta_12_filt_highD=Delta_12./1000; % microV to miliV, difference between lead 1 and
      lead 2

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Expected simulated sine
freq=10;
A=0.5; % mV, amplitude
dt=0; % s, nog aanpassen
S=A*sin(freq*2*pi*(tD-dt));

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Finding the equilibrium of the signal
min1=min(Delta_12_filt_highD(5000:10000));
max1=max(Delta_12_filt_highD(5000:10000));
dA=(min1+max1)/2; % calculates the equilibrium stance
Delta_12_filt_highD=Delta_12_filt_highD-dA;

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Find local peaks
M1=movmean(Delta_12_filt_highD, 15);
M2=bandpass(M1, [freq-1 freq+1], SampleFreq); % removing faster noisy frequencies
[pks1, locs1]=findpeaks(M2, 'MinPeakDistance', 0.8*(1/freq)); % finding the peaks of
      the mean
[pks3, locs3]=findpeaks(Delta_12_filt_highD, 'MinPeakDistance', 0.8*(SampleFreq/freq)
      ); % finding the peaks of the original signal

```

```

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Finding period of the signal
for i=1:(length(locs3)-1)
    Diff(i)=locs3(i+1)-locs3(i); % determines the distance between peaks
end
Nmeas=mean(Diff); % mean distance between the peaks in # samples
Tmeas=Nmeas*(1/SampleFreq);
fmeas=1/Tmeas; % returns the measured frequency of the signal

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Expected simulated sine
A=0.5; % mV, amplitude
dt=0; % s, nog aanpassen
S=A*sin(fmeas*2*pi*(tD-dt)); % Fitting a sine to the signal.

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Finding the phase shift between expected sine and
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% measured signal
[pks2, locs2]=findpeaks(S);
B=2*pi*fmeas;

if locs2(1)>=locs1(1)
    dt=tD(locs2(1))-tD(locs1(1));
    SD=A*sin(B*(tD+dt));
elseif locs2(1)<=locs1(1)
    dt=tD(locs1(1))-tD(locs2(1));
    SD=A*sin(B*(tD-dt));
else
    SD=0;
end

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Creating the overlay plot and RMSE value
Nmeas1=floor(Nmeas); % floored average # samples between peaks
CROSS=zeros(100,100); % With CROSS the columns corresponds to the single periodes:
    from peak n till peak n+1, the rows corresponds to the measurement values. This
    matrix is set before hand because every period has different length.
SDT=SD'; % transposing the sine to calculate RMSE value later

figure()
hold on
for i=1:1:(length(pks3)+1-floor(Nmeas/4)) %1/4 from the period is subtracted from the
    peaks, so the value at equilibrium is found
    A(i)=(locs3(i+1)-floor(Nmeas/4))-(locs3(i)-floor(Nmeas/4))+1; % delivers the
    length of the i'th period
    CROSS(1:A(i), i)=Delta_12_filt_highD((locs3(i)-floor(Nmeas/4)):(locs3(i+1)-floor(
        Nmeas/4))); % fills the cross matrix.
    RMSE(i,1)=rmse(CROSS(1:A(i),i), SDT(1:A(i),1)); % calculates the RMSE value for
    every column with a different length.
    plot(tD(1,1:A(i)), CROSS(1:A(i),i))
end

RMSE_tot_D=mean(RMSE);

for i=1:100
    MEAN(i)=mean(nonzeros(CROSS(i,:))); % calculating the mean of every row, without
    taking into account the zeros that are in their because not all columns have
    equal length
    STD(i)=std(nonzeros(CROSS(i,:)));
end

```



```

M3=movmean(MEAN, 17);
% A=max(M3)?
Amp=0.5;
S2=Amp*sin(fmeas*2*pi*(tD)); % the sine that will be in the overlay plot
plot(tD(1:100), MEAN, 'r', 'LineWidth',3) % mean of all the measured periods, -dA to
    correct for the offset.
plot(tD(1:100), S2(1:100), 'g', 'LineWidth',3) % expected sine
title('Textile electrode + conductive gel @ 10 Hz')
xlabel('Time [s]')
ylabel('Potential [mV]')
xlim([0 0.1])
ylim([-3 3])

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Calculating the 95% confidence interval
Upper=MEAN+STD.*1.98;
Lower=MEAN-STD.*1.98;

plot(tD(1:100), Upper, 'b', 'LineWidth',2)
plot(tD(1:100), Lower, 'b', 'LineWidth',2)

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Frequency plot of the signal
fftSignal = fft(Delta_12_filt_highD);
fftSignal = fftshift(fftSignal);

dF = SampleFreq/length(PolyCurves(:,1));
f = -SampleFreq/2:dF:SampleFreq/2-dF;

figure()
hold on
plot(f, abs(fftSignal), 'b')
title('Textile electrode + conductive gel @ 10 Hz')
xlabel('Frequency [Hz]')
ylabel('Magnitude')
ylim([0 5000])

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Plotting the total signal and all the relevant points
figure()
hold on
plot(tD, Delta_12_filt_highD, 'b')
plot(tD(locs1), pks1, 'k*')
plot(tD(locs2), pks2, 'm*')
plot(tD, SD, 'r')
plot(tD(locs3), pks3, 'y*')
plot(tD, M2, 'g')
legend('Measured signal', 'Peaks of bandpassed moving mean', 'Peaks of the expected
    sine (without phase shift)', 'Expected sine (phase shifted)', 'Peaks of the
    measured signal', 'Bandpassed moving mean of the measured signal')

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Plotting in the subplot
figure(13)
hold on
subplot(4,1,4)
hold on
plot(tD-0.06, SD, 'r')
plot(tD-0.06, Delta_12_filt_highD, 'b')
xlim([0.1 2.1])
ylim([-2 2])
xlabel('Time [s]')

```

```
ylabel('Potential [mV]')
title('Textile electrode + gel')
legend('Simulated signal', 'Measured signal')

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%% Returning overall STD
STD(isnan(STD))=0;
STD_all_D=mean(nonzeros(STD));
```

APPENDIX D

FIRST VERSION OF THE PROTOTYPE

A. Design and manufacturing

First, a Dipha-16 header with interchangeable wires was created (Figure 14 and 15). The Dipha-16 amplifier comes with various headers that can be clicked into place, accommodating three, six, or sixteen cables. In this application, the three-lead headers were utilized. The cables originating from a standard 3 lead Dipha-16 head were formed into a black female connector, in which pin headers could be inserted. These pin headers could be attached to a cable and an electrode.

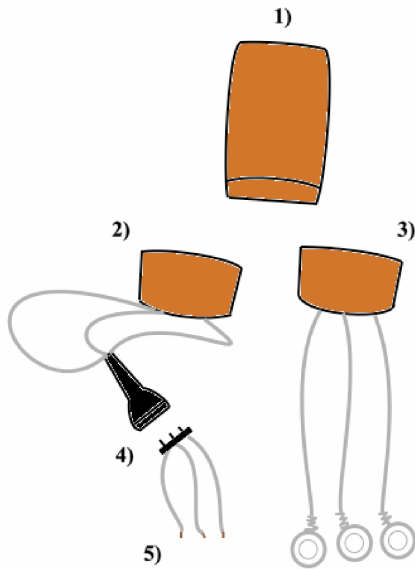


Figure 14: 1) shows a schematic drawing of the Dipha-16 amplifier without a header connected to the amplifier. 2) The customized header: pin headers can be inserted into the black female connection (4). These pin headers are connected to cables, to which prototyped electrodes are connected (5). In this way, prototypes can be changed without needing to create a new Dipha-16 header over and over again. 3) Current situation: a header that is internally connected to three cables and their snap connectors can be clicked on the standard Ag/AgCl electrodes.

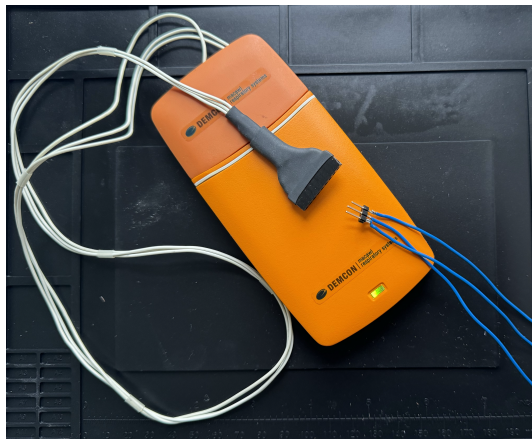


Figure 15: Customized Dipha-16 header that uses pin headers to connect the cables to the Dipha-16.

The design of the first prototype followed from literature. A literature search was conducted to find what material would be most suitable to create an electrode to measure dEMG in preterm infants [29]. To find answer to this question, the literature search was divided into four subquestions:

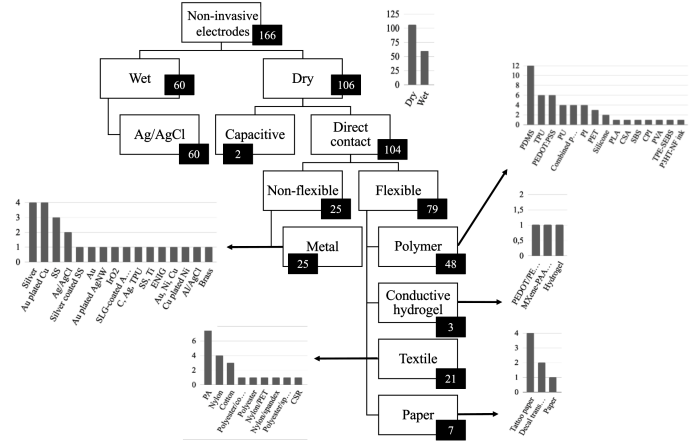


Figure 16: Hierarchically organized electrode categories and electrode materials. The frequency various electrode categories were described with, was shown within the small black rectangles next to the name of the electrode category. Arrows point towards the bar chart corresponding to a certain electrode category. The bar chart shows the frequency electrode materials are reported with, within that specific electrode category. Since not all electrode categories were described equally often, the y-axes of the bar charts differ in maximum value for each electrode category.

1) What electrodes are described in literature to measure non-invasive EMG in any application in humans and what are their advantages, disadvantages and properties regarding electrode material, SNR and size?:

In subquestion 1 electrodes that were retrieved from literature were divided into various subcategories and organized into a hierarchical table (Figure 16). 100% of the wet electrodes was an Ag/AgCl electrode. These electrodes are the state-of-the-art to measure non-invasive dEMG in humans. However, they are reported to be thick and big and they are not suitable for long-term monitoring due to gel dehydration.

Within the category of dry electrodes, a distinction was made between capacitive and direct contact electrodes. The capacitive electrodes were considered to be unsuitable to measure dEMG in preterm infants due to their susceptibility to subject movement and low SNR.

Dry direct contact electrodes were divided into flexible (textile, polymer, paper, conductive hydrogel) electrodes and non-flexible (metal) electrodes. Due to their rigidity and thus low skin conformance, dry direct contact metal electrodes were considered to be unsuitable to measure dEMG in preterm infants.

Within the flexible category, textile-, polymer-, paper-based electrodes and conductive hydrogels were analysed. Little to no information was found regarding the attaching and detaching process when using paper electrodes. Conductive hydrogels are new, complex, expensive materials, which makes them less suitable than more easily available alternatives.

However, certain polymers and textile electrodes have the

advantage of being comfortable and soft to the skin in addition to being readily available. Thus, when considering the advantages and disadvantages of various electrode categories, dry polymer and textile electrodes have greatest potential to be most suitable to measure dEMG within preterm infants.

When comparing SNR, SNR/area and $1/\text{thickness}$ for each electrode category, no significant differences were found between electrode categories regarding SNR and SNR/area. However, it was found that textile electrodes are significantly thinner (mean $1/\text{thickness}$: 5,28 (CI: 4,49 - 6,06) mm^{-1}) than the Ag/AgCl electrode (mean $1/\text{thickness}$: 0,25 mm^{-1}). Therefore, when considering advantages, disadvantages, SNR, SNR/area and thickness, textile electrodes were considered to be most suitable to measure dEMG in preterm infants.

2) What electrodes are described in literature to measure dEMG in humans?:

From subquestion 2 it followed that 100% of the electrodes used to measure non-invasive dEMG in humans is an Ag/AgCl electrode.

3) What electrodes are currently being used to measure non-invasive electrocardiogram (ECG) or non-invasive sEMG in preterm infants?:

Subquestion 3 showed that various methods could be used to measure ECG or EMG in preterm infants, such as smart mattresses, sensor belts and surface electrodes. Smart mattresses were not considered to be suitable to measure dEMG, due to the positioning of the diaphragm relative to the sensor and their susceptibility to subject movement. Sensor belts comprise a relatively large area of the infant's skin, which is undesired. Within the category of surface electrodes, 100% was reported to be an Ag/AgCl electrode. Most of these neonatal electrodes were pre-wired, making them less thick than the conventional Ag/AgCl electrode using snap connection. The absence of a snap connection makes them a suitable option to measure dEMG in preterm infants. However, no details regarding thickness and SNR of these pre-wired Ag/AgCl electrodes were found and the wires that are used are unshielded.

4) What electrodes can be used to measure dEMG in preterm infants?:

In the end, Ag/AgCl electrodes are currently the state-of-the-art method to measure non-invasive dEMG of the diaphragm. However, textile electrodes are soft to the skin, do not suffer from gel-dehydration, are thinner than the Ag/AgCl electrode, whilst showing no significant differences in SNR or SNR/Area. In addition, polymer electrodes could be suitable as well, but there no significant differences were found regarding thickness between polymer electrodes (mean $1/\text{thickness}$: 37,1 (CI: -104 - 178) mm^{-1}) and the Ag/AgCl electrode (mean $1/\text{thickness}$: 0,25 mm^{-1}). Therefore, it was concluded after the literature study that textile electrodes, and potentially polymer electrodes, are most suitable to measure dEMG in preterm infants.

So, as a first exploratory experiment, two options (Figure 17) for a sensing surface were created:

- A polymer: thermoplastic polyurethane (TPU) screen-printed with silver ink.
- A textile: a conductive fabric (Shieldit Super [28]) that can be ironed on cotton.

Of these two options, the textile option was the one having most advantages: its sensing surface had the lowest resistance, it was the easiest to manufacture and the softest to the skin. Therefore, this material was used as a basis for a first textile prototype.

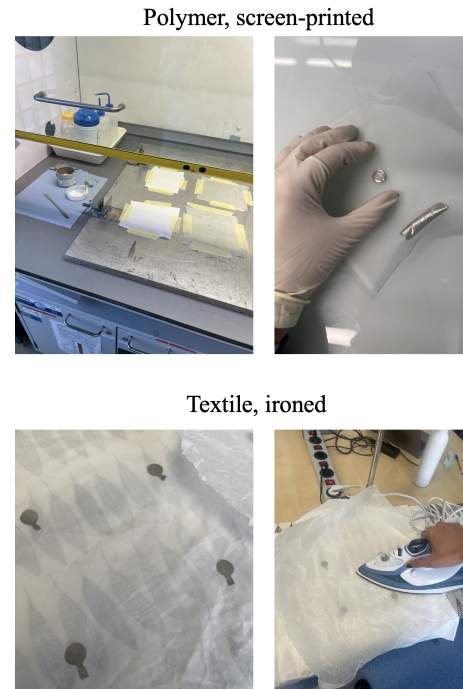


Figure 17: Photos showing the manufacturing process of a polymer, silver ink screen-printed sensing surface (top) and a conductive textile ironed onto cotton (bottom).

Finally, the first prototype was manufactured (Figure 18) by ironing Shieldit Super iron-on conductive fabric [27, 28] onto cotton. A circle with the same diameter as the sensing surface of the Ag/AgCl electrode (e.g. 7,6 mm) was created and a rectangular shape was added to the bottom to create a "spoon-like" shape. Later, the rectangular shape was used to attach the cable to the conductive fabric.

After the Shieldit was ironed onto the cotton, the cable was attached to the conductive surface. A technique as described in [7] was used. First, the end of the cable was formed into a loop. This loop was attached to the Shieldit conductive fabric by hand-stitching the loop onto the surface using conductive thread. A loop was used, so the cable would not be able to slide through the stitching when the cable is pulled. Lastly, the cotton was cut into the right shape using a fabric scissors and a piece of heat shrinking tube was used to cover up the electrode-cable connection.

B. Assessment of the prototype

Whether the prototype met the design requirements was assessed according to Table 1.

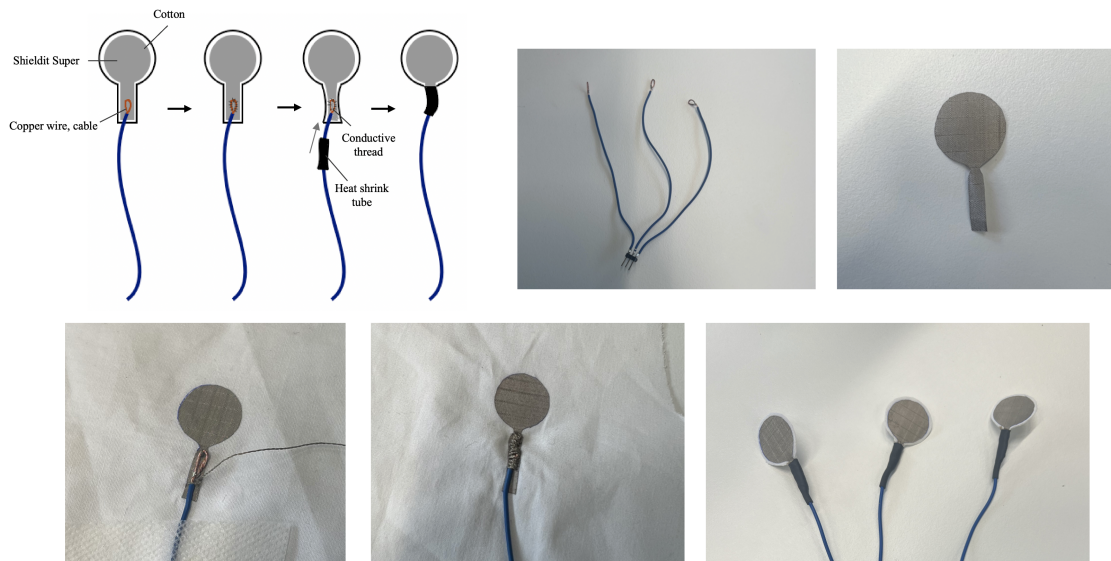


Figure 18: Manufacturing process of the first version of the prototype. First the ends of the cables were formed into loops. Then, the Shieldit fabric was cut into the desired "spoon-like" shape and ironed onto cotton. The loop was connected to the conductive fabric by hand-stitching using conductive thread and finally heat shrinking tube was used to cover the electrode-cable connection.

Electrode:

first, properties of the electrode itself were assessed. The prototype was considered to be breathable and biocompatible, since Shieldit conductive fabric has already been used in wearable sensors [18], and cotton is a common fabric in clothing and is known to be breathable. When considering costs, a piece of cotton costs $\text{€}5,95/\text{m}^2$ [13] and a piece of Shieldit Super costs $\text{€}91,60/\text{m}^2$ [28]. The textile electrode has a total surface of $4,5e^{-5}$ (circle) + $3,0e^{-5}$ (rectangle) = $7,5e^{-5}\text{m}^2$. Therefore, the costs for the electrode itself will be around $7,5e^{-5} * (91,60 + 5,95) = \text{€}0,01$. A piece of heat shrinking tube will cost approximately $\text{€}0,05$ [4]. If this is added to the price, the costs of one prototype would be $\text{€}0,06$. With or without heat shrinking tube, the costs remain below the price for one Ag/AgCl electrode (e.g. $\text{€}0,68$). The cable was not included within this calculation, since this is not included in the price for the Ag/AgCl electrode itself [14].

Since cadmium, mercury, chromium, PBB, PBDE, DEHP, BBP, DBP and DIBP were not used within the prototype, the prototype was considered to be RoHS compliant.

The prototype used three leads, connected to the Dipha-16 and the thickness of the electrode was only 1 mm (< 8 mm threshold). The diameter was set to 7,6 mm so that when comparing the prototype to the Ag/AgCl electrode, diameter could not be of influence on signal quality. Investigating the effect of a reduced diameter is of later interest.

In the end, all requirements regarding the electrode itself were met.

Electrode-cable connection:

The electrode-cable connection was established by using conductive thread (thickness without heat shrinking tube: 2 mm, thickness including heat shrinking tube: 3 mm) and not a snap

connection. However, when measuring the resistance of the electrode-cable connection, a resistance of approximately $900\text{ M}\Omega$ was found. This resulted in a prototype that was unable to measure any signal. When opening the shrinking tube, it was discovered that the conductive wire had started to fray, resulting in the loss of the electrode-cable connection, which was the start of the first iteration cycle.

C. First Iteration Cycle

Thus, in a second version of the prototype the electrode-cable connection needed to be improved. In order to make the right decisions, various options to connect cable to electrode were designed and assessed during Test A, of which the results were shown in Table 2.

This table showed that the connection using three small metal clips, a copper clamping tube, a big metal clip, woven copper thread through the conductive fabric and the thin conductive thread were the connections that could endure the highest force. Resistance values for conductive thread varied strongly, hence the $\pm$ sign. In addition, using conductive thread did not work in the first version of the prototype, so this method was excluded as an option to connect the cable to the conductive fabric. Of the options that were left, the woven copper wire was the most flexible and the thinnest (2 mm). Therefore, for a new prototype, weaving copper thread from the cable directly through the conductive fabric was chosen as the method to establish the electrode-cable connection.



	1	2	3	4	5	6	7	8	9	10	11	12
	3x small clips	Copper clamping tube	Big clip	Woven	Conductive thread (thin)	Conductive thread (thick)	Paperclip	Zipper ending	Press stud	Crocodile clip	Copper tape	Ag/AgCl
Resistance before [Ω]	0,4	0,4	0,4	1,3	+/- 4,4	+/- 1,5	+/-	0,2	0,2	0,3	0,4	6,7
Force [N]												
0,0981												
0,1962												
0,981												
12,26												
19,62												
24,53												
29,43												
49,05												
Resistance after [Ω]	0,8	1,1	0,5	0,5	+/- 1,9							6,7

Table 2: A table showing various methods to connect the cable to the conductive fabric. For each connection the electrical resistance of the connection before the pulling test was noted. Then, if a electrode-cable connection could withstand a certain pulling force, the square was colored in green. If the connection would break, the square was colored in red. If a connection could endure all pulling forces without breaking, the electrical resistance of the electrode-cable connection after the pulling test was noted as well. A +/- sign was noted if a resistance value varied strongly.

APPENDIX E

SECOND VERSION OF THE PROTOTYPE

A. Manufacturing method

The manufacturing process of this prototype was started in the same manner as the first version: by ironing Shieldit Super conductive fabric onto cotton. Then, a needle was used to pierce holes through the fabric in alternating positions as shown in Figure 19. The next step was to weave copper wire from the cable through the holes in the fabric and finally folding the fabric around the cable and covering it with heat shrinking tube.

B. Assessment of the prototype

Electrode:

The manufacturing process of the electrode itself remained unchanged and still met all the requirements that were set.

Electrode-cable connection:

As can be seen in Table 2, the woven cable-connection can resist pulling forces up to 49,05 N whilst maintaining good conductivity. The electrode-cable connection had a thickness without heat shrinking tube of 2 mm and a thickness including heat shrinking tube of 3 mm. All the design requirements regarding electrode-cable connection were met.

Skin-electrode interface:

Next, it was assessed whether the prototype met the requirements set for the skin-electrode interface. To adhere the textile electrode to the skin, various methods were designed. In the first option, the prototype was fixated directly onto the skin using a band-aid. This band-aid had already been used in medical applications and therefore was considered to be biocompatible. The second option was to use conductive adhesive gel [20], which is sold by Merkala [20] and is a certified product. Therefore, this option was considered to be biocompatible as well. The last option was to use conductive adhesive pads [1] to adhere the prototype to the skin. These pads are currently used in massaging devices or ab muscle training equipment and therefore considered to be biocompatible. In addition, no sharp edges were designed in the prototype. Lastly, to define impedance of the prototype and to verify whether the electrode did not fall off the skin spontaneously, Test B was performed. Impedance values were generated by the Natus software (Figure 20), from which Table 3 could be created:

	Average impedance [kΩ], CI (lower - upper)	
	Without skin prep	With skin prep
Ag/AgCl	85,6 (69,6 - 101,6)	70,6 (60,2 - 81)
Prototype + band aid	247,9 (-46,1 - 541,9)	174,3 (29,3 - 319,3)
Prototype + conductive pad	159,5 (147,3 - 171,7)	120,2 (106,2 - 134,2)
Prototype + conductive adhesive gel	142,1 (121,5 - 162,7)	81,4 (71 - 91,8)

Table 3: Impedance measurements in kΩ for the Ag/AgCl electrode and the prototype using various adhesion methods. The table shows the average value and between brackets the 95% confidence interval.

Raw impedance measurements were shown in Table 4. From Table 3 it could be derived that every version of the prototype had a higher skin-electrode impedance than the standard

Ag/AgCl electrode. Without skin preparation, the prototype adhered to the skin using a band-aid had highest impedance, with a large confidence interval. The prototype using a conductive pad and the prototype using conductive adhesive gel were in between the impedance of the Ag/AgCl electrode and the prototype using a band-aid. No clear difference in impedance could be found between conductive pad and conductive gel.

In addition, it could be seen from the table that after skin preparation all the impedances decreased, with the largest decrease in the impedances of the prototype combined with the band aid and prototype combined with conductive adhesive gel. Other than that, the differences in impedance between the various electrodes decreased as well. In the end, the prototype combined with a band-aid still had the highest impedance, followed by the prototype combined with a conductive pad, the prototype combined with conductive adhesive gel and lastly the standard Ag/AgCl electrode.

Even though measurements showed the large reduction in impedance when the skin was scrubbed, it could not be used in a final design of the prototype. The skin of the infants at the NICU is very delicate and can easily be damaged by a skin preparation scrub. Therefore, in following tests, the prototype was always adhered to the skin without preparing the skin.

Cable:

A standard shielded cable was used and was connected to the Dipha-16 by the customized Dipha header. Three leads were used to create a bipolar configuration.

Overall:

Finally, to test whether the prototype was correctly shielded and to assess signal quality of the prototype combined with every adhesion method, Test C was performed. During Test C (Figure 21) a copper plate was connected to the Fluke simulator. Electrodes were placed on the copper plate using one of three options: a band-aid, conductive pad and conductive adhesive gel. For each configuration, every period was layed over one another which was visualized in Figure 22. The RMSE and SD values were noted in Table 5.

Table 5 showed that the textile prototype in combination with a band-aid had the lowest RMSE value, followed by the Ag/AgCl electrode and the other two prototypes. Measurements performed using Ag/AgCl electrode had the lowest SD, followed by the textile prototypes, of which the textile prototype in combination with a band-aid had the lowest SD. In the frequency plot it could be seen that the magnitude of the 10 Hz signal in the Ag/AgCl electrode is smaller than the magnitude of the 10 Hz signal in the prototype combined with the band-aid. Unfortunately, amplitudes at frequencies regarding noise are higher in the textile electrode as well.

Even though the impedance of the prototype using a band-aid to adhere the prototype to the skin was the highest, this did not influence the measured signal due to the high input impedance of the Dipha-16 amplifier. Therefore, a band-aid was the selected method to adhere the textile prototype to the skin.

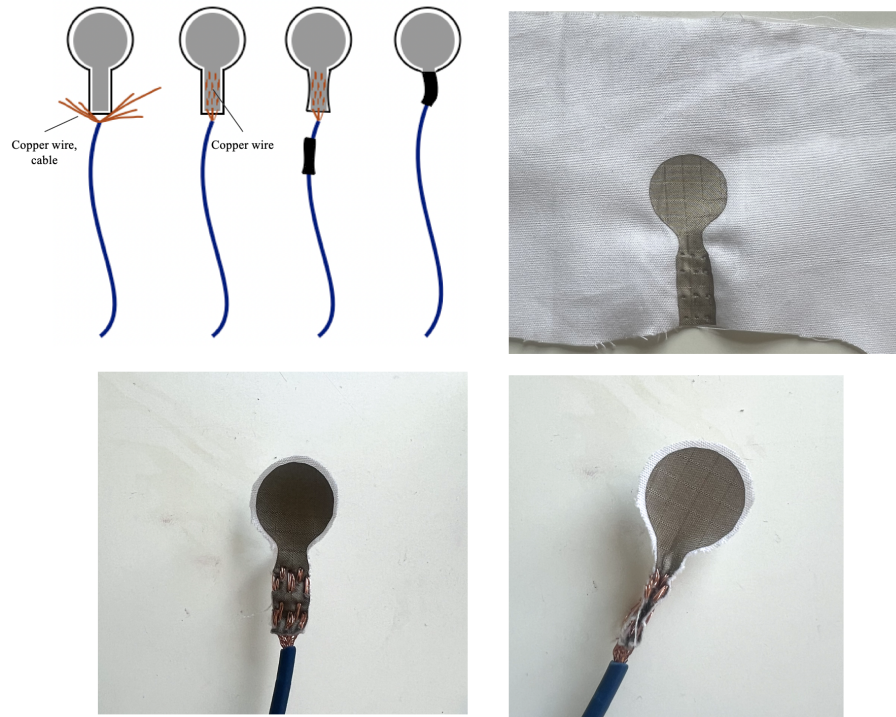


Figure 19: Manufacturing process of the second version of the prototype where the copper wire of the cable is woven through the Shieldit Super conductive fabric.

Without skin preparation												
	Ag/agel			Prototype + band aid			Prototype + conductive pad			Prototype + conductive adhesive gel		
	Earth	Silver cup	Prototype	Earth	Silver cup	Prototype	Earth	Silver cup	Prototype	Earth	Silver cup	Prototype
Lower arm 1	26,0	52,2	71,6	28,3	56,8	509,3	32,7	65,2	168,1	31,9	63,8	150,6
Lower arm 2	31,6	63,3	90,0	31,7	63,3	206,1	31,9	63,8	152,1	31,8	63,7	145,9
Lower arm 3	32,0	64,1	86,5	32,2	64,3	171,2	32,1	64,3	160,2	32,1	64,2	151,2
Lower arm 4	32,3	64,6	90,0	32,3	64,6	186,8	32,4	64,9	155,4	32,6	65,1	134,9
Lower arm 5	32,5	65,0	89,9	32,6	65,1	166,0	32,6	65,3	161,6	32,7	65,5	127,9
Average	30,9	61,8	85,6	31,4	62,8	247,9	32,3	64,7	159,5	32,2	64,5	142,1
Standard deviation	2,7	5,4	8,0	1,8	3,4	147,0	0,3	0,6	6,1	0,4	0,8	10,3
Coefficient of variation			0,09			0,59			0,04			0,07
With skin preparation												
	Ag/agel			Prototype + band aid			Prototype + conductive pad			Prototype + conductive adhesive gel		
	Earth	Silver cup	Prototype	Earth	Silver cup	Prototype	Earth	Silver cup	Prototype	Earth	Silver cup	Prototype
Lower arm 1	32,0	65,2	65,3	32,5	65,0	96,3	32,8	65,6	120,2	32,8	65,6	76,6
Lower arm 2	32,9	65,7	66,9	32,8	65,6	112,0	32,6	66,0	112,8	33,1	66,1	82,8
Lower arm 3	32,9	65,9	71,8	32,8	65,8	244,0	32,9	65,8	117,7	33,2	66,0	88,3
Lower arm 4	32,8	65,7	78,6	32,7	65,7	252,7	32,9	65,7	131,8	32,9	65,9	83,6
Lower arm 5	32,8	65,7	70,3	32,8	65,7	166,6	32,9	65,7	118,6	32,9	65,7	75,7
Average	32,7	65,6	70,6	32,7	65,6	174,3	32,8	65,8	120,2	33,0	65,9	81,4
Standard deviation	0,4	0,3	5,2	0,1	0,3	72,5	0,1	0,2	7,0	0,2	0,2	5,2
Coefficient of variation			0,073			0,416			0,059			0,064

Table 4: Impedance measurements in $k\Omega$

	RMSE [mV]	SD [mV]
Ag/AgCl	0,600	0,032
Textile + band-aid	0,151	0,113
Textile + pad	0,712	0,438
Textile + gel	0,736	0,200

Table 5: Table showing RMSE and SD values for the prototype using various adhesion methods and the Ag/AgCl electrode

C. Second iteration cycle

Unfortunately, as can be viewed from Figure 22, a 50 Hz peak still occurred in the frequency plot of the textile electrode even though a shielded cable was already used. Therefore, a next iteration cycle was started to try to make the electrode itself shielded as well.



Figure 20: Impedance measurement testing set-up using the Natus Nicolet AT2+6 amplifier and the built-in impedance test. The black wire was connected to the ground electrode, the red wire to a silver cup electrode and the green wire was connected to the electrode (prototype or Ag/AgCl) of interest.

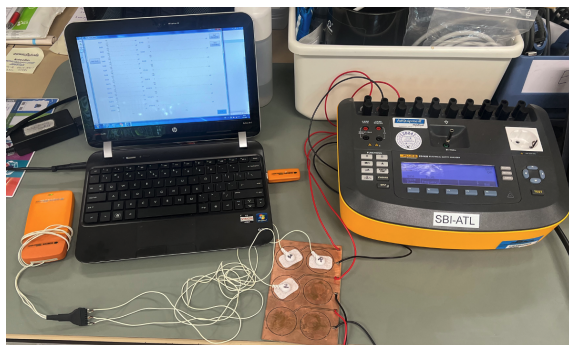


Figure 21: Testing set-up that simulated a sinusoidal signal. On the left, a computer running Polybench, connected to the Dipsa-16 amplifier. On the right, the Fluke simulator simulating the 10 Hz sinusoidal signal was placed. In the middle lies the copper plate with the electrodes attached to it.

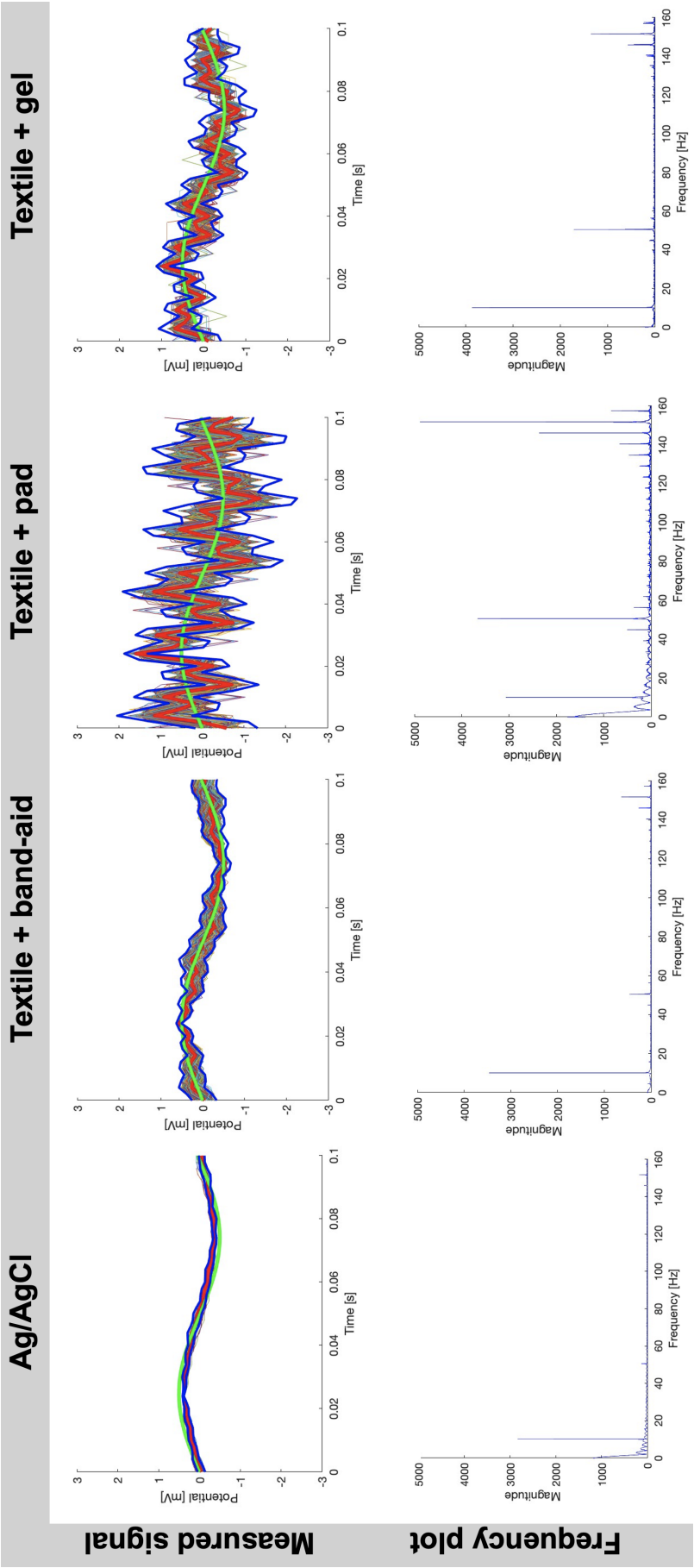


Figure 22: Graphs showing measured signals and frequency plots for the Ag/AgCl electrode and the second version of the textile prototype, combined with three different adhesion methods. Above: all the periods plotted over each other to define the RMSE value and the SD of the measured signals. The thick red line shows the mean of the signal. The thick blue lines show the upper and lower boundary of the 95% confidence interval and the green thick line shows the input sine. Bottom: frequency plots.

APPENDIX F FINAL PROTOTYPE

A. Design and manufacture

The inside of a shielded cable (Figure 23) consists of the copper wire of the cable, a PVC insulation (white), a spiral shield and a PVC jacket (black) [2].

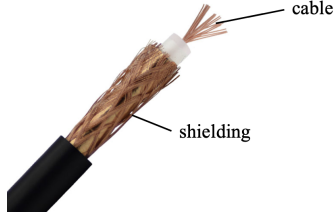


Figure 23: A shielded cable [2]. The black outer PVC layer is shown, followed by the network of copper shielding. Inside the shielding another white PVC insulation and within that insulation lies the copper cable that measures the signal of interest.

The spiral shield creates a Faraday's cage, so everything within the cage (the copper wire from the cable) is protected from electromagnetic interference.

First drawings and photos during the manufacturing of the electrode were included in Figure 25. First, two pieces of Shieldit conductive fabric were ironed onto two separate pieces of cotton. Then, the spiral shield around the cable was split from the cable itself. The copper wire of the cable was connected to one piece of Shieldit/cotton and the spiral shield was woven into another piece of Shieldit/cotton. Then, the two pieces of fabric were placed back-to-back against each other with an insulating layer of TPU between them. By connecting Shieldit/cotton to the spiral shield and placing it back-to-back against the sensing surface, it was aimed to protect the electrode itself from electromagnetic interference as well.

When considering costs, a piece of cotton costs $\text{€}5,95/\text{m}^2$ [13] and a piece of Shieldit Super costs $\text{€}91,60/\text{m}^2$ [28]. The textile electrode has a total surface of $4,5e^{-5}$ (circle) + $3,0e^{-5}$ (rectangle) = $7,5e^{-5}\text{m}^2$. Therefore, the costs for the two Shieldit/cotton sheets will be around $2 * 7,5e^{-5} * (91,60 + 5,95) = \text{€}0,02$. TPU costs $\text{€}157,10/\text{m}^2$ [21], so the total costs for the TPU layer were $7,5e^{-5} * 157,10 = \text{€}0,01$. A piece of heat shrinking tube will cost approximately $\text{€}0,05$ [4]. If this is added to the price, the costs of one prototype would be $\text{€}0,08$.

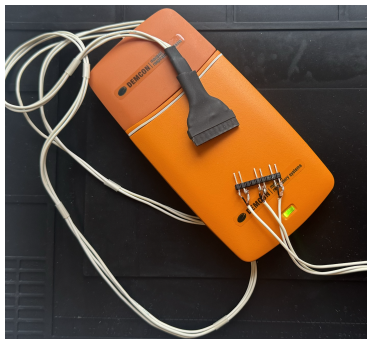


Figure 24: Customized Dipha-16 header when using shielded cables. Each cable is connected to two pin headers: one connected to the inner cable and one connected to the spiral shielding.

On the other end of the cable, the spiral shield and the cable needed to be connected to separate pin headers. The pin header of the spiral shield could not touch the pin header of the copper wire of the corresponding lead. Therefore, extra inputs in the Dipha-16 header were created (6 in total, 2 per lead) so that the pin headers could not touch one another (Figure 24). Finally, the cable connection was covered using heat shrink tube.

B. Assessment of the prototype

In addition to the topics discussed within the results of this paper, a figure was added that shows the measured 10 Hz signal of the final prototype in comparison to the Ag/AgCl electrode (Figure 26). Unfortunately, the frequency plot in this figure already showed that the design did not work and a 50 Hz peak still occurs within the frequency plot.

Also, a Harris profile was added to visualize how well both electrodes met the design requirements that were set.

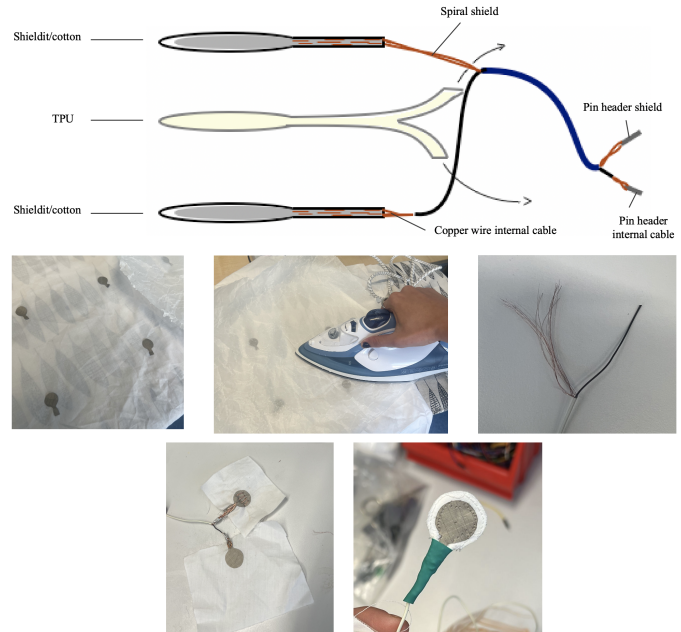


Figure 25: Manufacturing process of the third version of the prototype. The spiral shield was separated from the inner cable. Shield and cable were connected to two different pieces of Shieldit/cotton. Lastly, heat shrink tube was used to cover the electrode-cable connection.

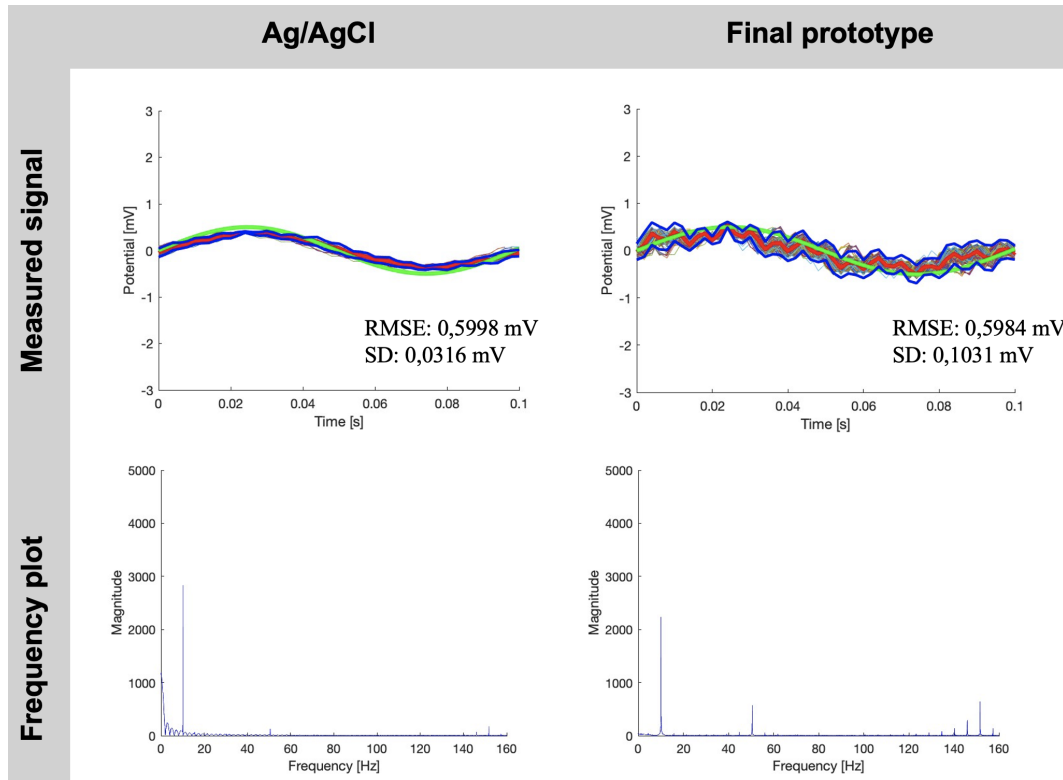


Figure 26: Above: shows the measured 10 Hz signal by the Ag/AgCl electrode (left) as well as the final prototype (right). The green line shows the expected input sine, the red thick line shows the mean of the signal and the thick blue lines show the boundaries of the 95% confidence interval of the measurements. Below: the frequency plots are shown.

		Ag/AgCl				Final design			
Requirement	Wish	-2	-1	1	2	-2	-1	1	2
Electrode									
Biocompatible	Breathable								
RoHS compliant									
	Low cost								
	Soft								
	Flexible								
Thin									
Small diameter									
Electrode-cable connection									
Not a snap									
	Flexible								
Thin									
Resistant to pulling forces									
Resistance as low as possible									
Skin-electrode interface									
Biocompatible									
No sharp edges									
	Dry								
	Low skin contact impedance								
Not falling of skin spontaneously									
Cable									
Connection to Dipha-16									
3 leads									
	Wireless								
Shielded									
Overall									
RMSE									
SD									
Total number of points									
Requirements only									
Requirements and wishes									

Figure 27: Harris profile of the final version of the prototype and the Ag/AgCl electrode

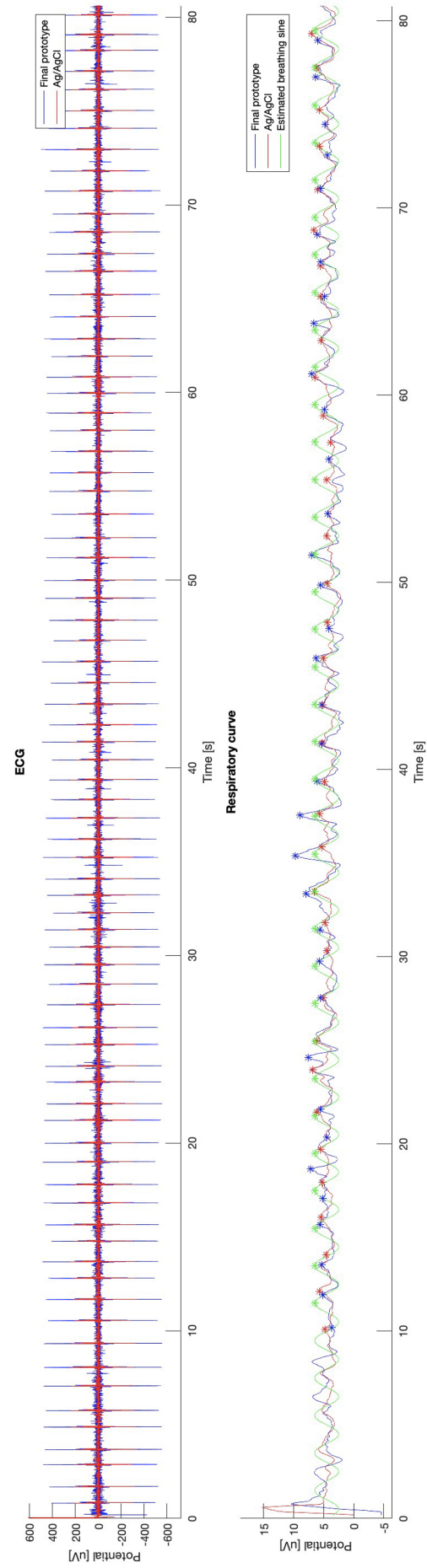


Figure 28: Total measured signal during the proof of principle test. Above the raw ECG is shown. Below: the breathing curve is extracted and the peaks are denoted using a *. Both electrodes measured 35 peaks in a time interval of 10 to 80 seconds, the same amount of peaks that can be counted in the simulated sinusoidal signal within that time.

