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groningen

Research Project

Faculty of Science and Engineering

Developing a pressure sensing sensorized sock based on piezoelectric pressure sensors, for performing gait analysis

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1 Abstract

In this research paper a sensorized sock prototype is developed for measuring gait parameters to indicate the disease Ataxia for patients. Ataxia is a degenerative disease where, due to the cerebellum losing its function, the patient has trouble walking. The sock developed in this research is based on piezoelectric sensors generating a voltage when being compressed. Here we show a prototype sensorized sock that is capable of measuring gait parameters, accompanied by an algorithm that can also give a preliminary Ataxia indication. Thin piezoelectric sensors based on a PVDF-TPU substrate were developed, during the research multiple iterations were used to optimize the design of the sensors. Sensors with a substrate of a higher thickness showed a more linear output, while the sensitivity compared to sensors with thinner substrates remained similar. The output for sensors with a high thickness was linear for at least the range from 60N to 180N. The sensitivity of sensors with interdigitated electrodes was approximately 0,052 Volts/Newton, while the sensitivity of sensors with normal electrodes was measured to be 0,031 Volts/Newton, this means the sensitivity of the interdigitated electrode sensors was 1,7 times higher compared to normal electrodes. The sensors showed good reliability for 1000 cycles of compression testing. It became apparent during testing of the prototypes that the sensors are prone to breaking under the force/friction generated by walking. Furthermore, it became clear that due to a lack of advancements in measuring piezoelectric voltage outputs, it is difficult to measure the voltage output accurately in a compact data acquisition system. Although it seems feasible to eventually create a sensorized sock based on piezoelectric sensors to measure gait parameters, this article concludes that it is better to focus efforts on smart insoles because of some of their characteristics such as their thickness and placement in a shoe.

Contents

2 List of Abbreviations

PVDF = Polyvinylidene fluoride

TPU = Thermoplastic polyurethane

IDE = Interdigitated electrode

DAQ = Data acquisition system

CU = Copper

DMF = Dimethylformamide

PVD = Physical vapor deposition

PDMS = Poly dimethylsiloxane

3D = Three dimensional

ADC = Analog to digital converter

3 Introduction

As we get more and more understanding of health problems, it is in our nature to help as many people as possible to solve these problems. A big part of improving health is monitoring and diagnosing of diseases. Currently, the field of smart wearable sensors is making great advancements as the technology behind these devices improves. Still examination and tests at a doctor are the most used methods of diagnosing diseases, although there are many people with mobility issues that would benefit from diagnosing and monitoring at their own homes. Patients with cerebellar ataxia experience deteriorating coordination, which also leads to a differentiated gate from a healthy person's gait. To diagnose and monitor patients the gait can be measured by using sensors that detect gait parameters. One commonly used method is to use smart insoles that can be placed in a patient's shoe. These insoles can measure parameters by the placement of pressure, acceleration, and other sensors. To enable testing without shoes and to reduce the material needed for the measurement of gait it is proposed to develop a sensing sock. In this research, a sensing sock based on piezoelectric sensors is developed in order to perform gait analyses on patients with ataxia.

4 Conceptual Research Design

4.1 Project context

4.1.1 Ataxia

Cerebellar ataxia is a degenerative disease that is caused by the cerebellum losing its function. From now on in the report by Ataxia is meant cerebellar Ataxia. Ataxia can be hereditary or it can be caused by damage to the cerebellum(i.e. impact, stroke). The most common symptoms related to ataxia are: Heart problems, Tremors, Lack of coordination, slurred speech, trouble swallowing, deteriorating motor skills, difficulty walking, gait abnormalities, and eye movement abnormalities (Ataxia.org, 2022). Ataxia can be subdivided into different types: Acute Ataxia, where suddenly a healthy patient shows symptoms of ataxia, usually children from 2-7 years old. Often there is an underlying condition such as head trauma, viruses, bacteria, certain vitamin deficiencies, and exposure to toxins. Intermittent Ataxia is a type of ataxia where symptoms come during episodes lasting minutes to hours. This type of ataxia is usually caused by gene mutations, it can also be hereditary. An episode can be triggered by emotional or physical stress. Chronic non-progressive Ataxia can be caused by a stroke or oxygen deficiency in the brain. The disease does not deteriorate over time. These non-progressive forms of ataxia are usually caused by (congenital) deformations of the cerebellum, the fourth ventricle, and a downward displacement of the cerebellar tonsils. Cerebellar progressive Ataxia is a disease where the conditions of the patient deteriorate over time. Often the cause of the disease is that there are inherited gene mutations present, which lead to a malfunctioning of the cerebellum. In children most often the cause is a cerebellar tumor (Pavone et al., 2017).

Very few of the hereditary types of ataxia can be cured, although some of the types with underlying symptoms can be cured for example restoring vitamin balances, etc. Currently, patients are helped in other ways with symptomatic treatment and supportive management in order to improve their quality of life (Pavone et al., 2017). One often-used method is physical therapy, where a patient has to perform exercises, this results in patients dealing with imbalance better and improving their gait. One of the most common and most noticeable symptoms of Ataxia is an abnormal gait. In section 4.1.3 features, and especially measurable/quantifiable features, of ataxic gait are described.

4.1.2 Pressure(Load) sensors

Currently, there are several types of sensors being used in insoles for measuring plantar pressure. In this section, a small description of each of these sensors is given. According to (Chen et al., 2022), the following are the six major categories of plantar pressure sensors.

4.1.2.1 Piezoresistive sensors

A piezoresistive sensor is a sensor that measures the change in resistance due to pressure. Piezoresistive materials have a change in resistance due to the pressure applied. This is measured and based on the magnitude of the change in resistance the pressure can be obtained. (Ge et al., 2021), made piezoresistive sensors with PEDOT:PSS on a graphene:PDMS wrinkled layer. The sensor was used for detecting knee extension. (Sengupta et al., 2019), made a piezoresistive sensor based on graphene-PDMS squeezable foam blocks. The sensor was used for gait detection and had a reliable output for at least 36000 cycles. The sensors were placed in a shoe sole and could detect heel strikes/offs, toe strikes/offs, and whether someone has flat feet or a higher arch.

4.1.2.2 Capacitive sensors

Capacitive sensors are based on the concept that the capacitance changes when the distance between two metal plates changes, where there is a dielectric between these plates. As the distance between the plates decreases, the capacitance increases. In (Dong et al., 2021), xylitol mixed with silicone rubber was used as a capacitive porous array. The sensor was used in a shoe sole and tested on different subjects. The sensor had a lifespan of at least 1000 cycles. Another paper fabricated capacitive sensors with PDMS sponge as the dielectric layer, PDMS/carbon black was used for the electrodes (Xia et al., 2021). The sensor remained robust for over 10000 cycles. Furthermore, the capacitive sensor was integrated into a shoe sole and heel strikes could be registered.

4.1.2.3 Inductive sensors

An alternating current is used to create a magnetic field, when the sensor is compressed the target gets closer to the magnetic field and using an LCR circuit the inductance is measured, which is increased when the target gets closer to the generated magnetic field. The article by (Nie et al., 2019) developed an inductive pressure sensor and tested it in the insole of a dancer. They used ferrite film, a fabric with 3D microstructures and a micro antenna to be able to detect an inductance change due to pressure exerted on the sensors. The sensors remained robust for over 20000 cycles, although at pressures higher than 20kPa the sensitivity decreased. This means it cannot accurately measure when higher loads are exerted on the sensor.

4.1.2.4 Optical pressure sensors

When pressure acts on an optical pressure sensor the light that is emitted is absorbed by a polymer differently, or the light reflects differently of the compressed surface. Because the sensor needs multiple components and is bulky it cannot be made very compact. According to (Chen et al., 2022), optical pressure sensors for gait monitoring show low robustness are expensive, and have a very complex design. For example, they name the article (Lincoln et al., 2012), where the sensor thickness is 5.5mm and reflective optical sensing was used, this article also highlights that the sensors are slightly temperature sensitive.

4.1.2.5 Air-pressure based sensors

Tubes with air are placed and when compressed the pressure inside the tubes increases and is measured using an air-pressure transducer. In order to measure the pressure on different locations on a foot multiple air bladders need to be placed under the foot (Chen et al., 2022). This leads to a bulky insole and makes walking uncomfortable. If further steps in miniaturization and optimization of the air-pressure based sensors is done they might become less bulky and more suitable for gait monitoring.

4.1.2.6 Piezoelectric sensors

In this section, the working mechanism of piezo-electric materials will be explained. Furthermore, current research on flexible piezo-electric substrate materials is further elaborated upon. Lastly, a detailed explanation is given of the working mechanism of PVDF as a piezo-electric substrate.
|Working mechanism| The piezo-electric effect is the ability of a material to generate a voltage, due to deformation of crystals in the material. Additionally, the reverse piezo-electric effect is the ability of a material to deform when a voltage is applied to the material. As in this research the piezo-electric material is needed for sensing, the reverse piezo-electric effect is not of interest. As can be seen in Figure 1, in its rest position a crystal has no potential difference. In Figure 1B and C a potential difference in the crystal is created by compression of the crystal. This potential difference creates an alternating current when establishing a circuit around the crystal. In practice this is done by adding electrodes with wires to the piezo-electric material. As the pressure on the crystal increases the deformation increases which results in an increase of charge generated.
|Flexible piezoelectric material| In this research piezoelectric sensors need to be in-

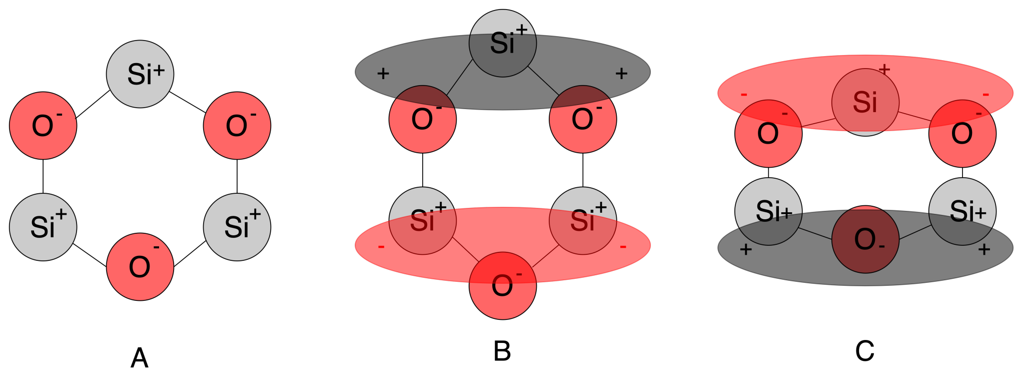


Figure 1: *Mechanism of creating a potential difference when deforming a crystal. [A] The crystal is in its neutral position, there is no potential difference. [B] The crystal is compressed from the sides, a potential difference is created [C] The crystal is compressed from the top, a potential difference with reverse poles compared to (B) is created*

tegrated into a sock, so the piezo-electric material must be flexible. According to (Sappati and Bhadra, 2018) "Extensive focus has been given in the last two decades to develop new piezoelectric compounds such as piezopolymers, piezo papers and lead free piezoelectric materials". One major piezo-polymer the article highlights is PVDF(Polyvinylidene fluoride). (Sappati and Bhadra, 2018) states that "PVDF is a special polymer which has many crystalline forms in which beta-phase creates a high piezoelectric effect". So in order to reach optimal piezoelectric properties in the flexible material PVDF, when obtaining the substrate for the sensor, the beta-phase of PVDF must be obtained. In the beta-phase the polymer chains are highly ordered, contrary to alpha-phase PVDF where the material has a random orientation. The highly ordered chains in the beta-phase improve the piezoelectric output because the dipoles are aligned, which results in a higher total potential difference. There are several methods to obtain the beta-phase of PVDF, for example: uni-axial stretching(Li et al., 2014), poling, and electrospinning. In uni-axial stretching the PVDF material is stretched during heating, for example in (Sencadas et al., 2006) it was stretched 5 times the original length at 80 degrees Celsius, to obtain a beta-phase content 0f 80%. During poling a voltage is applied to align the electric dipole moment, for example (Hartono et al., 2016) sandwiched PVDF foil between two electrodes, and at a temperature of 40 degrees a voltage of 20kV was applied, this resulted in a maximum beta-phase content of 83%. During electrospinning the PVDF substrate is prepared by using a PVDF solution, the solution is extruded from a nozzle during which an electric field is applied between the nozzle and the deposition surface (Lim et al., 2015). Because of the electric field, the solution reaches the deposition surface as elongated fibers. (Lim et al., 2015) obtained a beta-phase content of over 90% using electrospinning.
The previously named sensors can all measure in such a way that you cannot determine all gait parameters. In order to be able to measure distance between steps additional sensors are needed in addition to the plantar pressure sensors. A combination of plantar pressure sensors and gyroscopes/accelerometers could be used

to also determine spatial gait parameters. In this research the focus is solely on piezoelectric plantar pressure sensors.

4.1.3 Gait

Gait is the manner in which a person walks. There are many different parameters to compare between people and to help decide how a person's gait can be improved. Sensors can be placed all over a person's lower limbs, for example: hip angle, knee angle, ankle angle, speed, amount of steps per minute, ground reaction force, etc can be measured. Since in this report the design is limited to piezo electric sensors, placed in a sock, not all gait parameters can be measured.
Multiple research papers have investigated differences in gait between ataxia and control patients. In one paper, (Buckley et al., 2018), data was collected from several previous study's comparing ataxia and control patient's gait parameters. (Buckley et al., 2018) made a distinction between several domains:

-Pace domain. Two parameters are compared in the pace domain, speed and cadence, both parameters show a reduction in pace in patients with ataxia compared to the control groups.

-Spatial domain. Stride length and step length are two parameters that are both reduced in patients with ataxia relative to control patients. Furthermore, patients with ataxia have an increased base width during walking.

-Temporal domain. Stride time and stride length are also significantly higher in ataxia patients compared to healthy controls.

-Gait cycle domain. In the gait cycle domain it becomes clear that both double limb support phase and stance phase constitute a larger portion of the gait cycle in ataxia patients, while at the same time the swing phase is reduced. This indicates that the patient has more contact with the ground to find support.

-Variability domain. Ataxia patients also show increased variability in their gait compared to the control groups. Step length, stride length, stride time and speed all vary more in ataxia patients.

Utilizing the data output generated by the sensorized sock, gait parameters can be extracted. These parameters could be displayed to the patient/healthcare personnel to help decide whether a patient does or does not have ataxia. Furthermore, these parameters could be used after a patient is diagnosed with ataxia to determine how the disease is regressing(Schmitz-Hübsch et al., 2006).
Because of the characteristics of the sock developed in this study, not all parameters in all cohorts can be obtained. For example, the pressure sensors cannot indicate distances, so parameters in the spatial domain are not available. The following parameters are chosen as indicators to give an indication of whether a patient does or does not have ataxia:

-Cadence
-Swing phase

-Stride time variability

Based on the data from (Buckley et al., 2018), for each parameter, a cut-off value is determined to indicate whether based on each individual parameter the patient might have ataxia or not. These values are implemented in the algorithm, and the algorithm gives an estimation whether the person does or does not have ataxia. An Ataxia indication for each of these gait parameters is given. For cadence, Ataxia is indicated for a cadence lower than 105 steps/minute. For swing phase, Ataxia is indicated for a swing-phase lower than 38%. For stride time variability, Ataxia is indicated when the variability exceeds 5%. (Stolze et al., 2002) explain how gait parameters are influenced by ataxia, "The cadence was significantly lower in CD(Cerebellar disease), and the stance phase and double limb support duration showed that more time was spent in contact with the ground."

4.1.4 Algorithms detecting gait patterns

In order to provide the patient and healthcare personnel with information based on the sensorized sock an algorithm is needed to collect, store and calculate data/-parameters. One aspect is the live display of the sensor output and the cadence. Another aspect is the calculation of gait parameters based on previously collected data. Furthermore, an algorithm could compute an indication of whether a person has ataxic gait or not based on threshold values of certain gait parameters, see section 4.1.3.

4.1.4.1 Live display of data

Measurement, computation, and display of data at the same time requires computing power, in order to prevent lag or the need for a lot of computing power the live data display should be optimized for efficiency. A computationally cheap method to calculate the frequency, and thereby the cadence, of a signal is the fast Fourier transform (Jaber and Massicotte, 2009). The fast Fourier transform is an efficient way of calculating the discrete Fourier transform. Based on the amplitude spectrum of the Fourier transform the dominant frequencies can be determined. Due to background frequencies that can be present in the signal the frequency determined from the amplitude spectrum should be within a range that includes normal walking and running frequencies.

4.1.4.2 Computing gait parameters

After the collection of the data, this data can be used to compute gait parameters of the subject. This can be done after the data collection has finished, or at certain intervals to have intermediate updates on gait parameters. When a pressure acts on a sensor a positive voltage signal is obtained, and when this pressure is released a negative signal is obtained. In order to detect when pressure is applied and released, thresholds can be set for each sensor to determine at which threshold a pressure event will be detected.

4.1.4.3 Thresholds for ataxia indication

Following the computation of the gait parameters, the cadence, swing phase, stance phase, and stride time variability can be used to indicate whether a patient might have ataxia. For each parameter that is used the values obtained from data are compared with known values of ataxic gait parameters.

4.1.5 Competition

An important part of studying the feasibility of a product is to determine the level of competition.
(Lehmann and Winer, 2008) determine four levels of competition. Each form of competition is defined below including examples for the product of interest here: A sensorized sock to measure gait parameters for patients with Ataxia.

Product form competition This competition is based on similar products. In this case similar products are other sensorized socks used for gait monitoring(of Ataxia patients).

Product category competition This is competition by products in the same category. These are products that fullfill the same basic objective, although they might reach that objective in a slightly different way. So in this category fall sensorized insoles, video-based gait monitoring, different types of wearable sensors(i.e. placed at the knee and hip to measure angles), watches or phones that include step counters.

Generic competition Generic competition comes from products that full fill the same basic need. In the case of this research, the generic competitors are doctors performing manual methods to determine if someone has Ataxia and how the disease progresses.

Budget competition Budget competition are products that fight for the same budget of a consumer, since the sensorized sock is for a specific target group and does not compete for the budget of a person, no budget competition is present.

At the time of this research, no sensorized socks specifically developed for patients with Ataxia exist. There are few sensorized socks developed for measuring gait parameters. (Amitrano et al., 2022) investigated the performance of Sensoria smart socks(sensoriafitness.com, 2023). In their research, it became clear that the sock was able to measure cadence with a bias of -0.0820 steps/min. The sock underestimated the stance phase duration, the average bias was -7.75%. Since it is necessary for measuring gait parameters regarding Ataxia to know the stance phase, the current Sensoria sock does not suffice as an instrument to measure Ataxia in patients.
(Lucangeli et al., 2022) created and analyzed the Sistine 2.0 sock with four force sensors placed on the sock, protected by cloth to prevent damage. The sistine 2.0 sock was not able to guarantee a reliable accuracy as reached by force plates, although it was able to give an indication of the balance trend.

A sock developed by (Tirosh et al., 2013) showed good results measuring gait parameters. There were slight errors in stride duration and stance duration of 1.6% and 3.8% respectively. The main issue regarding the sock developed by (Tirosh et al., 2013) is that the bottom of the sock is sewn to a neoprene shape of a foot, which adds a thickness of an additional few mm to the bottom of the sock, which could influence the gait pattern and cause discomfort during walking.

Multiple sensorized insoles have been developed(Saidani et al., 2018), insoles compared to socks have more thickness and are therefore more facile to fabricate, furthermore larger/thicker sensors can be included in the insole. The main disadvantage of insoles compared to socks is that insoles have to be placed inside a shoe for walking, while wearing a shoe the foot support is changed and the gait pattern can change(Wegener et al., 2011).
Video based gait monitoring is used as well for measuring gait patterns and even a warning system has been developed based on in home video analysis of gait patterns(Grobe-Einsler et al., 2021). The main disadvantage of video based gait monitoring is that multiple camera's are used and the camera's are stationary, which makes the area of analysis very limited.
Other wearable sensors used for gait monitoring are placed at the knee, hip and foot to determine the angles of the joints. This is a complicated system, needing pants with sensors included for comfortable and easy measurement at home. Compared to wearing socks, which are small and easy to put on, wearing sensorized pants is more intrusive and limiting for the patient(Kadirvelu et al., 2023).

4.2 System description

The system of interest in this research is a smart sock system that has the aim of being used in gait monitoring of ataxia patients. The following components are part of the smart sock system:
The system exists of two sensors integrated into a sock in such a way that the sock can be worn without a shoe and be used during walking inside.
In order to be able to interpret the data for the end consumer the voltage output of the sensors has to be converted into information such as cadence and temporal gait parameters. So the data acquired must be stored, an algorithm is needed to transform the data and an interface can be used to display the final data.
Power source, External memory, Microcontroller, and transceiver are not taken into account during this project due to time limitations. Instead during the project existing data acquisition devices are used such as a DAQ and oscilloscope.
The whole smart sock system will have efficacy in the monitoring of ataxia patients. And this will help with making healthcare more accessible and efficient.

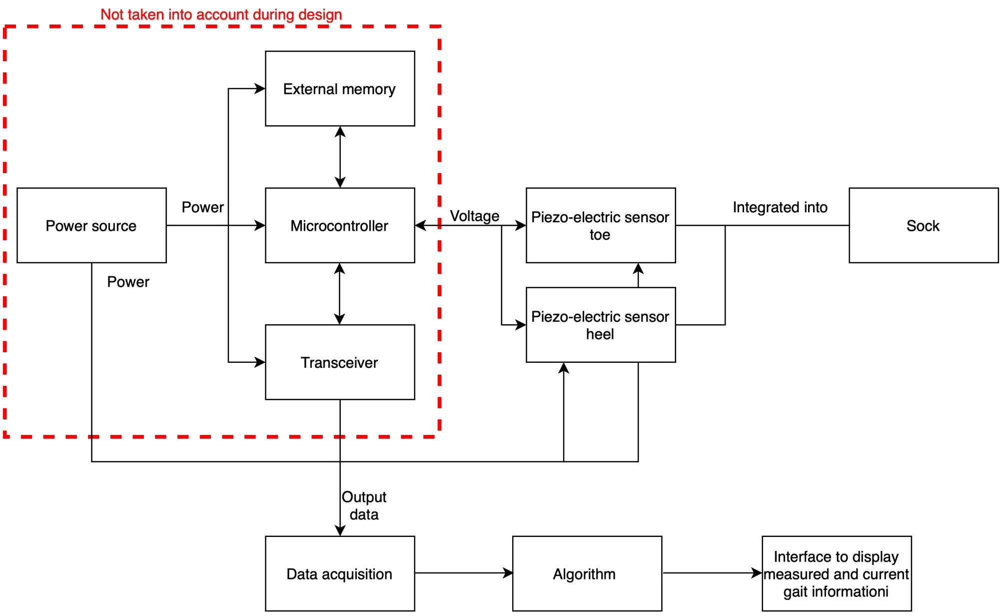


Figure 2: *System description: the red dotted defines components not taken into account in this research, furthermore the system exists of two piezoelectric sensors, integrated into a sock, and the digital part where the acquired data is transformed into gait information.*

4.3 Problem statement

Currently it is difficult to monitor patients with ataxia and their walking patterns, they usually have to come to the hospital or another healthcare facility to be able to get their gait measured. Measuring at home with sensors in shoe soles or socks can help with that, but it is expensive and shoe soles have been developed to a certain extent but are less optimal than socks, since wearing insoles also influences the gait which could have a positive or negative influence on a person's gait. Pressure sensors in socks are still in the beginning stages of development. This leads to the following problem statement:

There is no cost-effective sock available that measures pressure based on piezoelectric sensors, to obtain gait parameters for ataxia patients

In the following section the potential stakeholders when this project is commercialized are highlighted. The main entities which have influence on the research project, or are affected directly by it are listed below. Furthermore the potential stakeholders for commercialization are named and some insights are given how they would influence the outcome of the final product.

4.4 Stakeholders

The main entities which influence the research project, or are affected directly by it are listed below. Furthermore, the potential stakeholders for commercialization are

named and some insights are given on how they would influence the outcome of the final product.
The following stakeholders are involved with the research project:

University of Groningen Universities are the entities that have as a goal to educate and to generate knowledge. This research can contribute to the knowledge base of the university and the project is also a form of education for me, the student.

Research group APE The research group advanced production engineering works on manufacturing processes, and mechanical and materials engineering. And will benefit from research and discoveries done inside their labs. Performing research will also contribute to the research group in a monetary way, because if more students are doing research in the labs more funds are available for the group.

Apart from the previously named stakeholders for the project, there are also potential future stakeholders when the sensorized sock is commercialized.

Consumers There are many potential users for a smart sock, for example athletes, people doing personal training, people with podiatry related problems, and elderly people. The main people that should be taken into account for this project are people with ataxia. Since people with ataxia already have difficulty walking their interest is to have a product that can help detect the severity of their disease without compromising their current ability to walk, possibly even improving current gait patterns.

Wearable technology companies Companies that are currently selling smart wearables, could be interested in selling a sensorized sock to diversify their product range. Furthermore, similar companies could become competitors when they also develop sensorized socks. There are many companies selling smart wearable and/or healthcare technology such as: Apple, Fitbit, and Garmin and also newer smaller companies such as Athos, Atlas tracking and Neurotech. According to (Ometov et al., 2021) "Wearable technology has a tremendous impact upon ICT industry, and smart wearables are expected to disrupt most personal and business sectors, such as the industrial, healthcare, and sports domains". This indicates there are many opportunities for wearable technology companies to grow and launch new smart wearable products.

Insurance companies/Dutch healthcare authority In order to get medicine or medical technology approved for the healthcare market, it must adhere to the standards set by the Dutch healthcare authority. The Dutch Healthcare Authority makes policy on which medicine and medical technology are reimbursed for everyone (In the Netherlands everyone needs to have basic health insurance). (Zorgautoriteit, 2022) Furthermore, for some medical technologies the insurance companies/healthcare providers can determine whether they reimburse the product or not.

Doctors When a smart sock is used for measuring data regarding Ataxia patients, doctors will also be involved. The data extracted from the sock must be according to their standards and they should be able to use the software accompanying the sock.

4.5 Goal statement

Based on the previous problem statement and the stakeholder analysis a goal statement is developed, the goal is to:

Develop a prototype sensorized sock, based on piezo-electric sensors, to measure gait parameters and determine the feasibility of developing a sensorized sock for commercial use

In order to achieve this goal, several sub-goals are stated to make the main goal more concrete.

1: *Develop, test and optimize piezo-electric sensors*

The first step to developing the prototype is to develop the piezo-electric sensors. After the initial sensor is developed efforts should be made to test the sensor and as a result of these tests the design of the sensors could be optimized.

2: *Determine characteristics such as sensitivity and reliability of the sensors*

In order to be able to compare the sensors to other piezo-electric sensors that have been developed the sensors must be characterized. Important aspects of the sensors are their reliability and sensitivity.

3: *Fabricate a sock with the piezo-electric sensors integrated*

After the sensors have been developed they should be integrated into a sock. There are design considerations that have to be made, and again tests have to be done to determine how the sensors operate when integrated into the sock.

4: *Develop an algorithm that can determine heel strikes/offs, and toe strikes/offs and which is able to determine gait parameters from this data*

When the sock is fabricated tests on the sock should be done and the voltage output obtained from the sock(s), should be used in an algorithm to determine whether the sock can be used to obtain gait parameters.

4.6 Research Approach and Methods

By using a methodological approach to the research and by using the design cycle(Wieringa, 2014), the research can be conducted in a more structured and organized way. In this research the design cycle by Wieringa is used, see Figure 3. The first step of the cycle is to identify the problem, stakeholders and goal. The next step is to design the artifact, which in this research is a sensorized sock. The final step of the design cycle is to validate whether the design accomplishes its goals and objectives or not. In this research, these three steps of the design cycle will all be conducted at least once. Within certain steps other cycles of improvement exist as well, for example the design and fabrication of the sensors should also include testing and redesign.

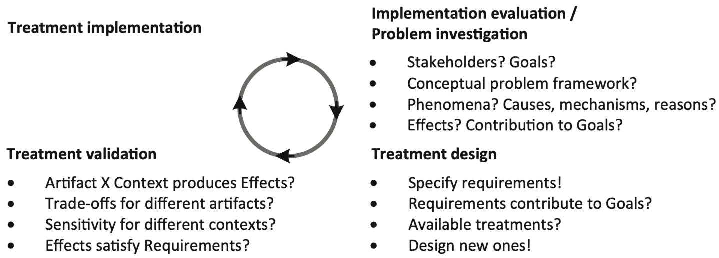


Figure 3: *Design Cycle; The Design Cycle constructed by (Wieringa, 2014) consists of the first three steps of the engineering cycle. The cycle can be followed multiple times, after one iteration the problem investigation step is extended with implementation evaluation.*

4.7 Research Questions

Based on the goal set for the research, some research questions are proposed which need to be answered to obtain the goal.
In order to be able to answer the first sub-goal the following sub-questions are asked:

How to develop piezo-electric sensors?

How to test and optimize piezo-electric sensors?

In order to reach the second sub-goal the following research question is asked:

How to characterize the piezo-electric sensors?

The following two sub-questions are asked to answer the third sub-goal:

How to fabricate the sensorized sock?

What are important factors to take into account regarding the structure of the sock?

To obtain the final sub-goal the following research question is asked:

How to develop an algorithm based on the voltage output of piezo-electric sensors?

To determine the feasibility of the sensorized sock, the next research question is asked:

Is it feasible to develop a sensorized sock based on piezo-electric sensors?

In the technical research design, the approach to answering the previous research questions is described.

5 Technical Research Design

5.1 Operationalization

After the finalization of the conceptual research design, the technical research design is started. First of all the sub-questions will be operationalized by describing how the questions will be answered.

How to develop piezo-electric sensors?

In order to determine how to develop piezo-electric sensors, a literature review is used to find suitable materials and compositions of sensors. Furthermore, assistance on fabrication methods by fellow students and researchers at the university lab is used. Also, design tools such as computer aided design or mathematical formulas can be used to design the piezo-electric sensors and their dimensions.

How to test and optimize piezo-electric sensors?

To determine the best way to test the sensors literature review, trial and error, and expert opinions are used. Firstly after developing the sensors experiments in the lab should be done to test if they are usable, further experiments should be used to specify points of improvement. Optimization of sensors will be done partly by deduction, to determine what are current faults and using logical reasoning and further literature survey how to improve the current designs.

How to characterize the piezo-electric sensors?

Characterization of sensors can generally be done in many different ways, using various types of equipment. Because the availability of equipment at the lab is limited, literature review and availability of the equipment, are used to determine which characterization methods are useful. Experiments will be performed using different types of equipment and lab equipment will be used for the characterization.

How to fabricate the sensorized sock?

In order to answer this question literature review will be used, and trial and error will be used as well, so trying a design and improving based on outcomes.

What are important factors to take into account regarding the structure of the sock?

In order to make the sock durable, the structure of the sock should be taken into account, this can partly be done using literature search, experiments , and trial and error.

How to develop an algorithm based on voltage output of piezo-electric sensors?

In order to develop an algorithm first the output of the sensors should be measured and taken into account. Using literature search similar types of algorithms can be found. Than using programming, trial and error and experiments the algorithm can be constructed and improved upon.

Is it feasible to develop a sensorized sock based on piezo-electric sensors?

To determine the feasibility a minor market analysis must be done using literature search, and the performance of the sock must be compared to similar products available.

6 Materials & Methods

6.1 Substrate fabrication

PVDF-TPU composite is used as a substrate for the sensor due to the piezo-electric properties of PVDF and the flexibility of TPU.
A solvent must be used to be able to electrospin the composite. In this case a solution with a ratio of DMF:Aceton of 6:4 was used. DMF is used due to its strong dissolving properties. Aceton is used to lower the boiling point of the solution.
According to (Shehata et al., 2022), the optimal piezo electric response is obtained when the PVDF:TPU ratio is 85:15, furthermore the study confirmed that the resulting substrate is flexible and stretchable. The tensile strength was 3.8 MPa and the breaking strain 82%. Even though for the application in this paper such large breaking strain is not needed, the optimal piezo electric properties of 15% TPU substrate makes it the most desirable option.
The solution was obtained by weighing 0.85 grams of PVDF, weighing 0.15 grams of TPU pellets, and stirring this for 12 hours at 60°C in a solution with 6ml of DMF and 4ml of Aceton. This solution was drawn up into a 10ml syringe. The syringe was placed in the NE300 electrospinner by Inovenso, the solution was electrospun on aluminum foil at a rate of 1ml/hour for 5 hours at room temperature. A flat collector slightly moving side to side was used to obtain a more uniform substrate.

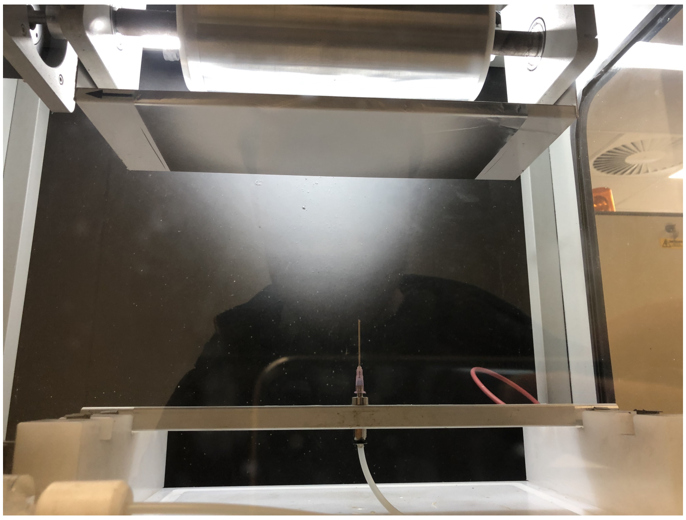


Figure 4: *Picture of the electrospinning setup(NE300 electrospinner by Inovenso), on the bottom side there is a nozzle attached to a syringe, the aluminum foil is above the nozzle and the plate to which the aluminum foil is attached moves from side to side*

6.2 Interdigitated electrode fabrication

Electron beam physical vapor deposition was used to deposit the gold on the outside of an adhesive plate seal. Gold was used as thin film electrode material because of its high conductivity and good mechanical properties during bending(Pawlak et al., 2017). A mask, see Figure 5, was designed in Solidworks software and was laser-cut from stainless steel by the university workshop. This mask was clamped to the substrate holder. The gold in the crucible was than vaporized by the electron beam and thereby the gold was deposited on the adhesive plate seal. The arms of the IDE's were designed to be 1mm wide, decreasing the width of the arms and thereby being able to have more arms would increase surface area and potentially increase the output of the sensors. Due to limitations of laser cutting for the stainless steel, the arms could not be decreased below 1mm.

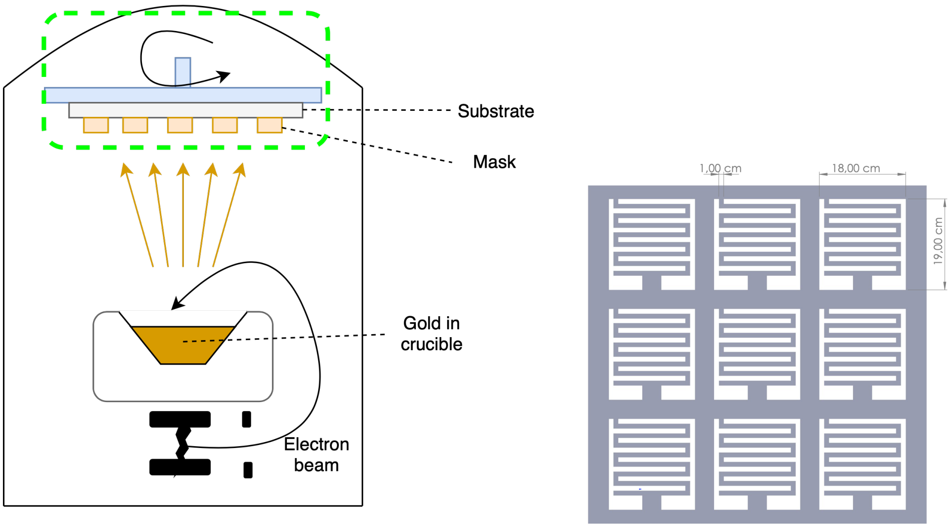


Figure 5: *[Left] Electron beam physical vapor deposition of gold IDE's, where the substrate is adhesive plate seal and the electron beam is aimed at the gold in the crucible to vaporize the gold. [Right] Mask used during the deposition to obtain the interdigitated electrode design. The width of the arms of the electrodes is 1mm.*

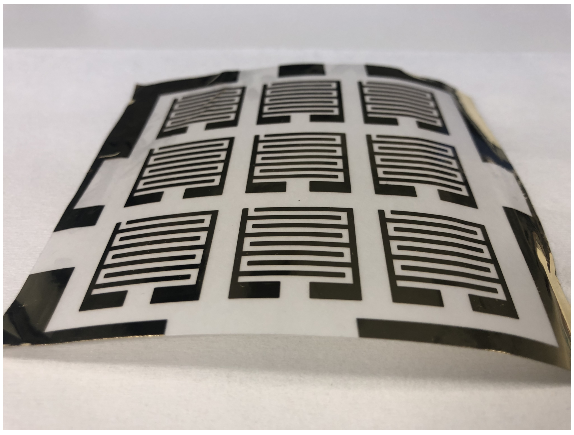


Figure 6: *Picture of gold interdigitated electrodes deposited on the adhesive foil*

6.3 Sensor development

The electrospun PVDF:TPU substrate and IDE's were used to fabricate the flexible piezo electric sensors. For the different sensors fabricated slightly different fabrication methods were used. This is due to the fact that the IDE process is more extensive. The process for fabricating the sensors with interdigitated electrodes is further elaborated on.
The simplified diagram in Figure 7, shows the fabrication steps of a sensor. Adhesive plate seals are used and the PVDF-TPU substrate is attached to the adhesive. The aluminum foil is removed and step 2 is obtained. Then the IDE's are placed on the substrate and the wires are attached to the electrodes. Lastly, the sensor is closed with a second adhesive plate seal. Both type of sensors obtained are displayed in Figure 8.

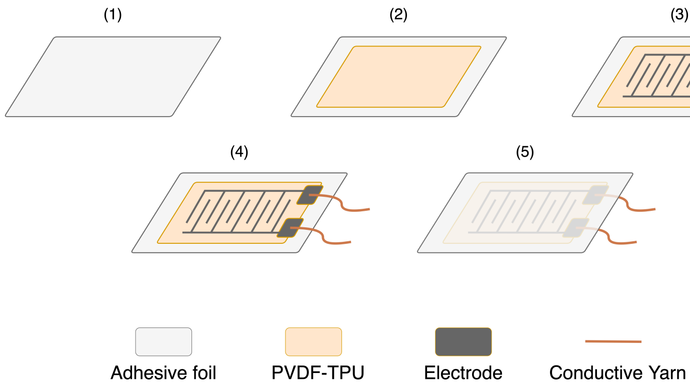


Figure 7: *1: Adhesive foil is used as packaging around the sensor; 2: PVDF-TPU foil is added on the adhesive foil; 3: IDE's are assembled on the; 4: Conductive yarn is attached to the IDE's; 5: The sensor is closed of*

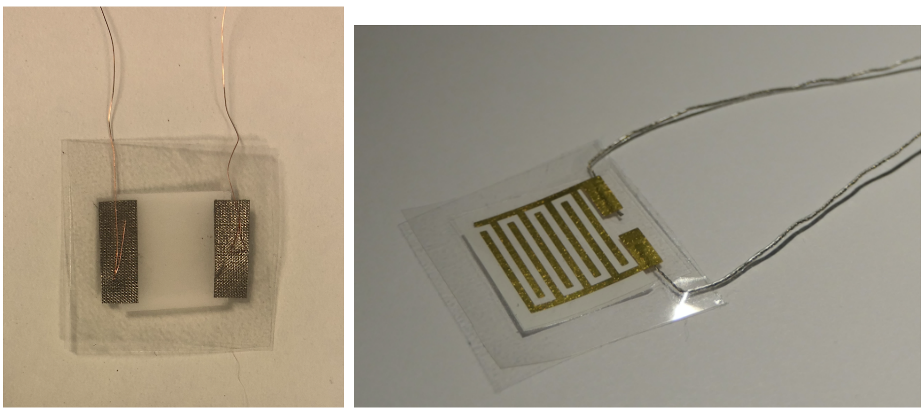


Figure 8: *[Left] Picture of initial sensor with two rectangular electrodes and copper wires [Right] Picture of sensor with IDE's*

6.4 Sensor testing

6.4.1 Characterisation

To determine the properties of the developed PVDF-TPU substrate, first of all the thickness of the substrate is measured, a thickness gauge 0-12.7mm by Neoteck was used. The resolution of the thickness gauge is 1µm.

6.4.2 Sensitivity

The sensitivity of the sensor was measured by comparing the peak-to-peak voltage output at different loads on the sensor. The voltage output is measured using a National instruments USB-6289 DAQ. In order to measure the actual force exerted in a controlled manner a Cellscale univert mechanical tester is used. This mechanical tester can apply loads at predetermined loads and frequencies.

6.4.3 Reliability

The reliability of the sensors is tested and the output of the sensor is compared for 1000 cycles.

6.4.4 Frequency response

The compression test equipment has limited frequency settings, so the frequency response of the sensor could not be exactly determined. The Cellscale univert mechanical tester was used to exert force at a frequency up to 1 Hz.

7 Results

In this section the results of the research are presented. First the characterisation of the PVDF-TPU substrate is expanded on. Next, the sensitivity and reliability of the sensors is demonstrated. Furthermore, the steps of integrating the sensors into a sock and problems that arise during this phase are further explained. After which the algorithm created in Matlab is further illustrated. Lastly, an overview of the results is given.

7.0.1 Characterisation

After electrospinning the PVDF-TPU on aluminum foil, some initial sensors were fabricated. The transfer of the substrate into the sensor posed some difficulties, because some sensors had uniform substrates while other parts of the substrate did not show good adhesion to the encapsulation of the sensor. This was observed as color differences in parts of the foil, which is caused by lesser thickness. And at certain areas no substrate was present after transferring the substrate to the sensor. These issues called for a longer electrospinning time, so the time was increased from electrospinning for 2 hours to electrospinning for 5 hours. To try and prevent further issues during fabrication steps the thickness of the newly electrospun substrate was measured. This showed major differences in thickness, the thickness profile can be seen as a dome shape, so the middle of the substrate has a significantly higher thickness than the edges. Multiple tests were conducted to determine the minimum thickness needed for proper adhesion of the substrate. An approximate minimum thickness of 30 µm is needed for proper transfer of the substrate to the sensor. Although even for higher thicknesses some inconsistency's remained during transfer of the substrate.

7.0.2 Confirm output is piezoelectric

To confirm that the output measured is indeed a piezo-electric signal and not a different type of signal created by for example electrode compression a test is done with an oscilloscope. In Figure 9, it can be seen that the output of the sensor is a typical piezo-electric output. When pressure is applied a spike in voltage is generated, which immediately settles back to the baseline. When pressure is released a negative spike is generated, which similarly returns to zero. On the left of Figure 9, there is a time interval between pressure and release, on the right there is no time between pressure and release, the sensor is "tapped". Noticeable is also that the spike on release of pressure is smaller than the spike created by the pressure.

7.0.3 Sensitivity

Measuring the sensitivity of the sensor can give some insight in the possibilities of using the sensors to determine not only if there is a load present, but to determine the magnitude of this load. So, for different sensors sensitivity plots were created, presenting the voltage output versus the force exerted on the sensor.
First of all, sensors with and without interdigitated electrodes are compared. Sec-

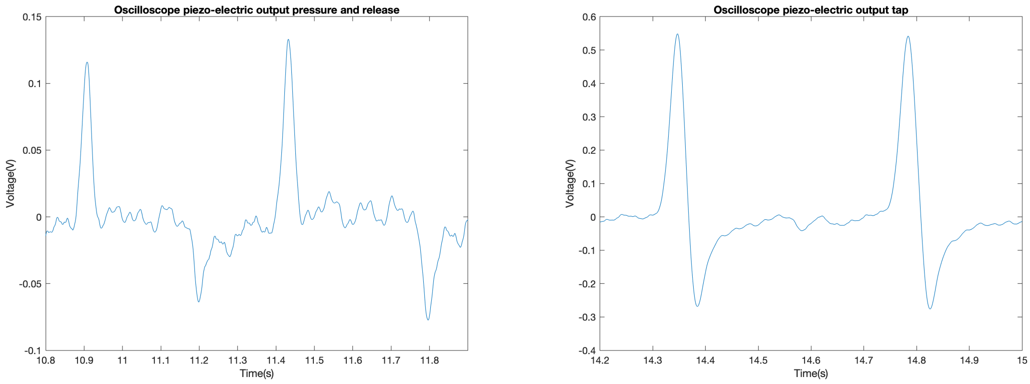


Figure 9: *[Left] The piezo-electric output displayed is pressure and a release a few tenths of a second later. [Right] The pressure is released immediately after being applied, so more a tap than a prolonged pressure*

ondly, the sensitivity of sensors with different substrate thickness are compared to determine if the output can be improved by increasing or decreasing the thickness of the substrate.

7.0.3.1 Normal electrodes versus Interdigitated electrodes

In order to improve the output of the piezo-electric sensors interdigitated electrodes can be used. Interdigitated electrodes increase the contact area between electrodes which is expected to increase the voltage output for the same force applied. In this section normal electrodes are compared to interdigitated electrodes. In Figure 10, the sensitivity plots of normal and interdigitated electrodes are shown. The voltage output at the different forces applied for the normal sensor and the IDE sensor are displayed in Appendix A and Appendix C.
Due to limitations in the measuring equipment, the DAQ can only measure voltage output between -10V and +10V, for higher force exerted the IDE voltage output could not be measured accurately. So, only for 20N,60N, and 100N the output was measured. It becomes clear that the output with interdigitated electrodes is significantly higher than without interdigitated electrodes. It is not clear whether the linear range of the sensor with interdigitated electrodes also starts between 20-60 Volts similar to the normal sensor, for the following calculation regarding the increase in sensitivity it is assumed it does not start at a higher load than 20N. The slope of both trend lines is compared. To calculate the sensitivity the following formula is used:

$$Sensitivity(Volts/Newton) = \frac{\Delta Voltage}{\Delta Force}$$

The sensitivity of the IDE sensor is approximately 0,052 Volts/Newton, while the sensitivity of the sensor with normal electrodes is approximately 0,031 Volts/Newton. Based on these values the sensitivity is approximately 1,7 times higher for an IDE sensor compared to a sensor with normal electrodes.

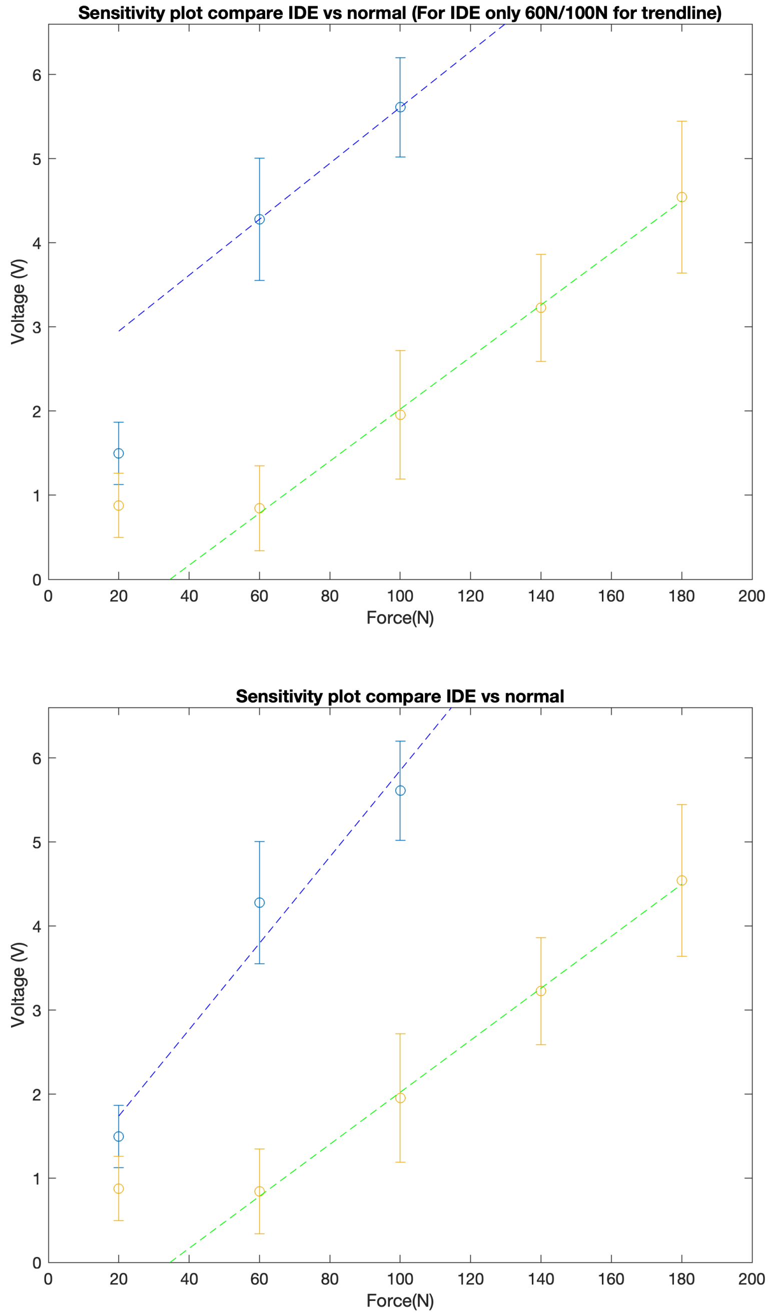
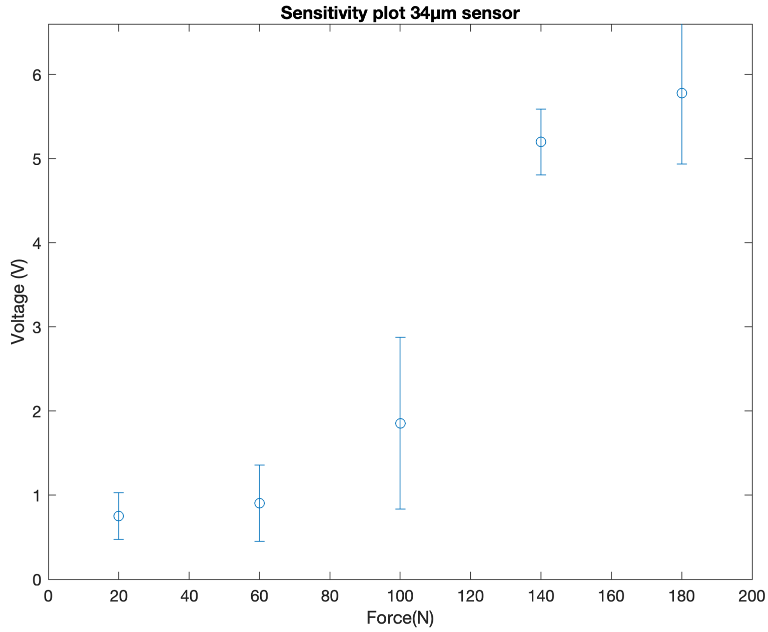


Figure 10: *Sensitivity plots of IDE(Blue) and normal electrode sensors(Yellow/Green). The sensitivity test was repeated 3 times for each force from 20N-180N. The plots include error bars with a 95% confidence interval and a trend line.*

7.0.3.2 Substrate thickness deviation influence on sensitivity curve

One parameter that can influence the piezo-electric output of a piezo-electric sensor is the thickness of the substrate used. In this section, the sensitivity plot of two different substrate thicknesses is compared. In order to compare the sensitivity the voltage output is measured for each sensor over a range of forces from 20N to 180N with a step size of 40N. For each magnitude of force the experiment was re-iterated 3 times and based on the data obtained sensitivity plots were created, with a 95% confidence that the actual value lies within the error bars. It can be seen in Figure 11 that both sensors have a positive correlation between force applied and output voltage. Furthermore, the plot of the low-thickness(34µm) sensor shows less linearity compared to the high-thickness(51µm) sensor. The high-thickness sensor especially shows good linearity when not taking into account the 20N force output of the sensor. This gives an indication that for the high-thickness substrate sensor the linear region starts between 20N and 60N and at least continues until 180N. From the voltage output of both sensors the, average sensitivity of both sensors is similar, this indicates that increasing the thickness of the substrate does not increase the sensitivity of the sensor. Although, on the other hand, it becomes clear from the plots that a higher thickness of the substrate does lead to a more linear voltage output. So, when using the piezo-electric output to not only determine if there is a force applied, but to determine the magnitude of the force a substrate thickness of 51µm is preferable compared to a substrate thickness of 34µm. The voltage output at the different forces applied for the normal sensor and the IDE sensor are displayed in Appendix A and Appendix B.



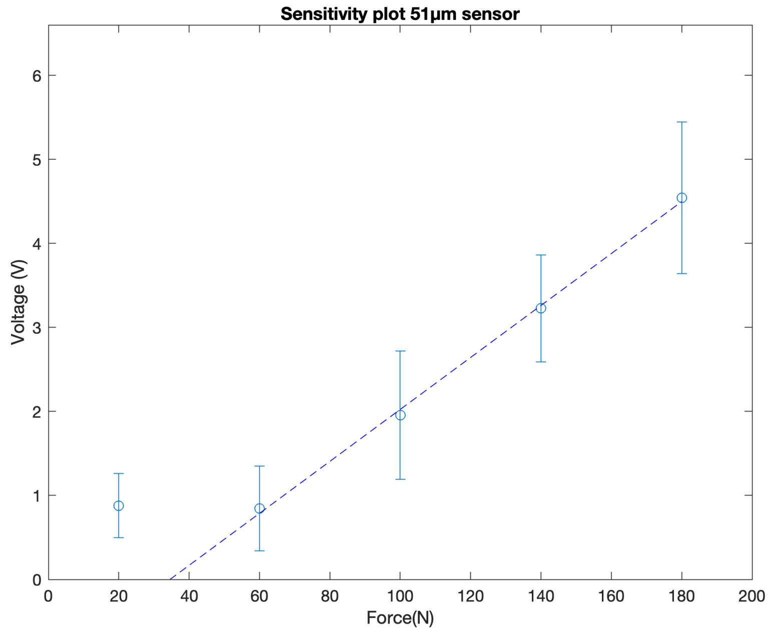


Figure 11: *Sensitivity plots of two sensors(34µm and 51µm). The x-axis shows the force and the y-axis displays the voltage output generated at each force. The sensitivity test was repeated 3 times for each force from 20N-180N. The plots include error bars with a 95% confidence interval and a trend line.*

7.0.4 Reliability

In order to determine if the sensor will remain operational after many cycles of load are applied, a reliability test is performed. The reliability of both normal electrode and interdigitated electrode sensors is compared.

7.0.4.1 Normal sensor reliability

In Figure 12 the reliability test of a high thickness sensor with a thickness of 51µm is displayed. It can be seen that after 1000 cycles the sensor still generates an output and the magnitude of the output does vary slightly, although overall it remains fairly constant. Furthermore, there is some drift that can be seen over the time of the experiment.

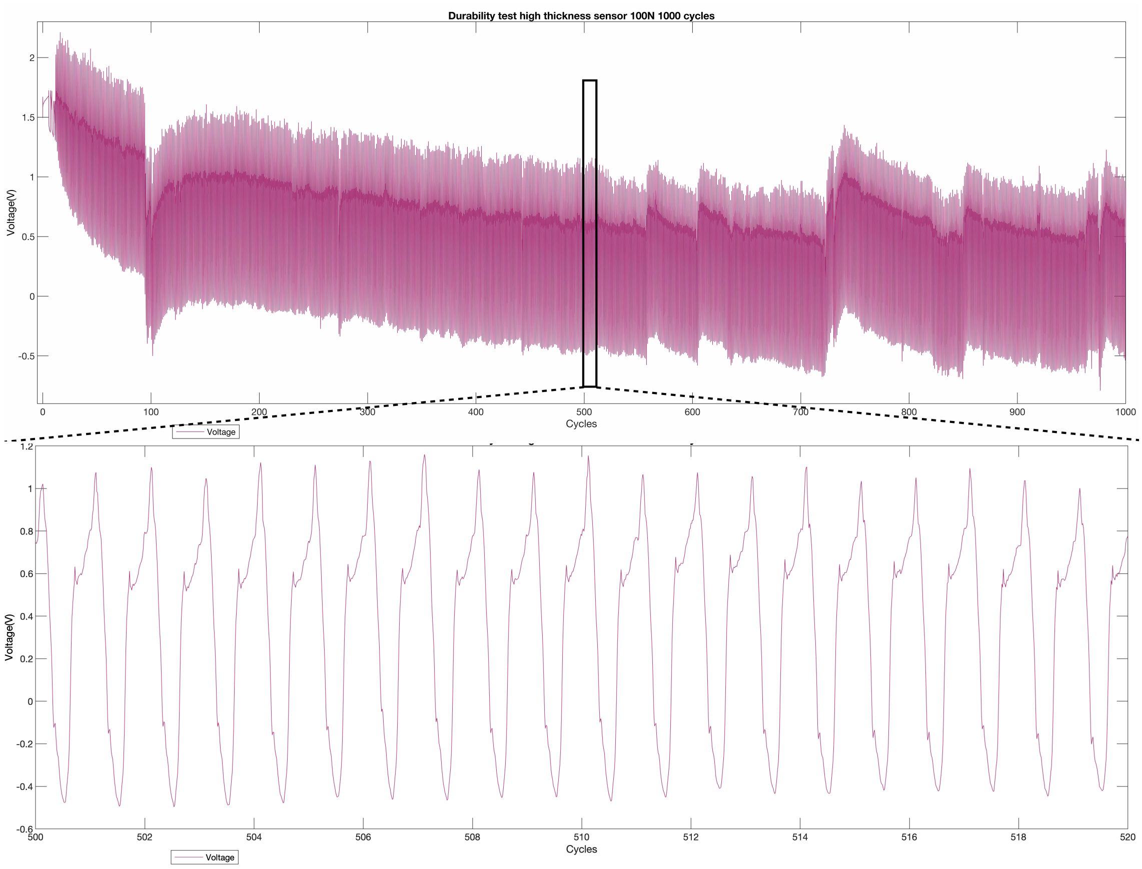


Figure 12: *Sensor with normal electrodes compressed for 1000 cycles with a load of 100N, with a frequency of 0,4Hz*

7.0.4.2 IDE sensor reliability

The interdigitated electrodes show a larger voltage output, even for 1000 cycles of compression. From Figure 13 it becomes clear that the output is less constant and uniform compared to sensors with normal electrodes. The main difference in testing is that with the test on interdigitated electrodes also pressure on the electrodes themselves is exerted, which leads to some large deviations in output.

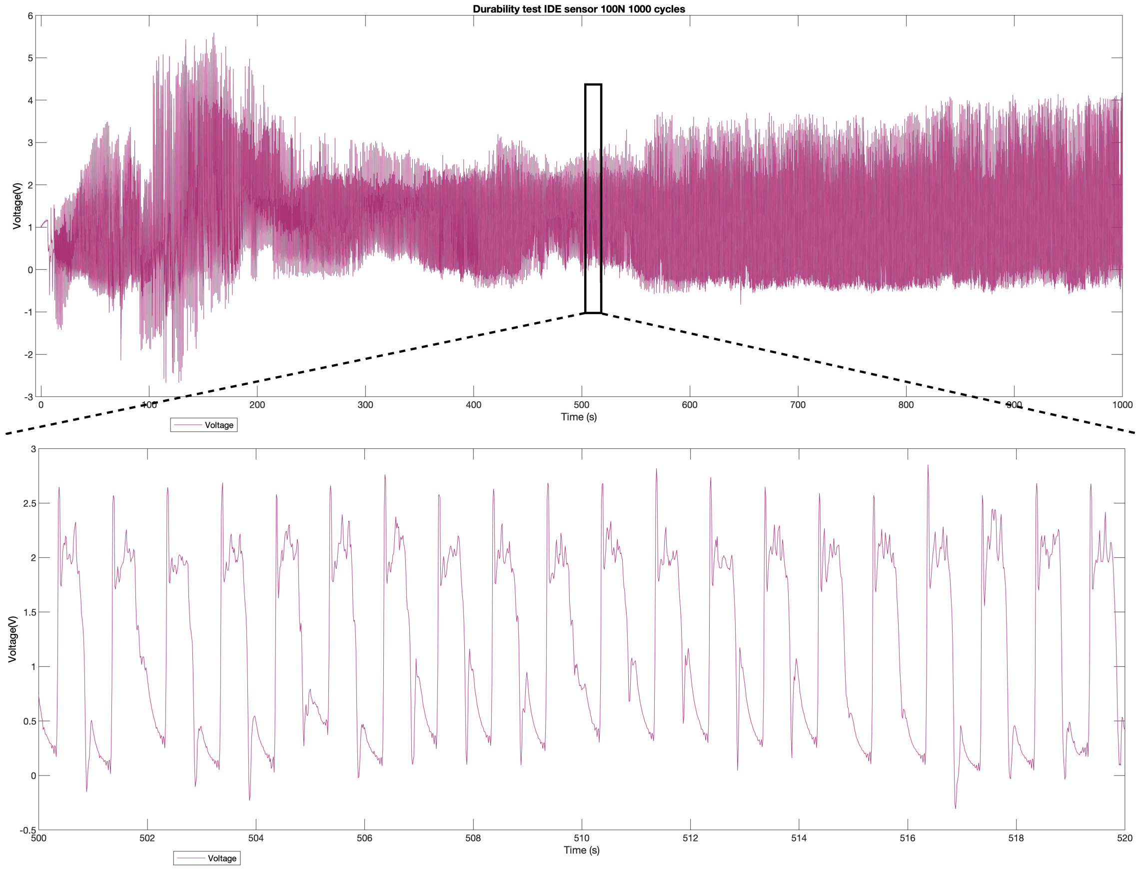


Figure 13: *Sensor with interdigitated electrodes compressed for 1000 cycles with a load of 100N, with a frequency of 0,4Hz*

7.1 Design and integration into a sock

The piezo-electric sensors are integrated into a sock. Firstly, an initial design was made based on the first iteration of the sensors. For this design one sensor was placed at the heel and one sensor was placed at the toe of the sock. The first design was made early on in the project to be able to determine based on this design what might need to be changed later on in the project. In Figure 14 the initial design is shown. It can be seen that the sensors are sewn in the sock, in order to be able to do this the sides of the sensors had to be perforated. Furthermore, the wires were attached at different places using sewing as well.

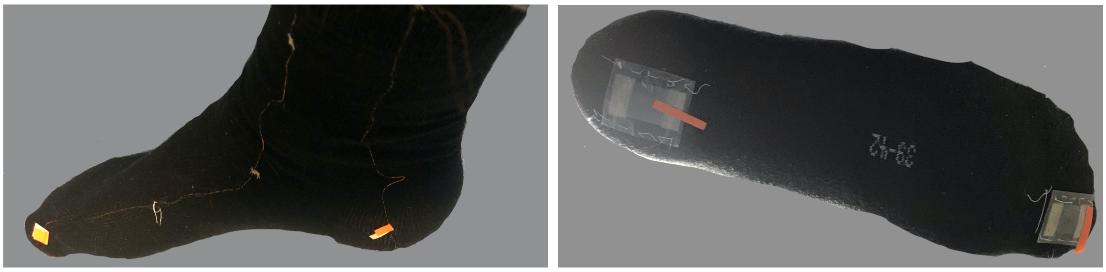


Figure 14: *Initial sock design, the sensors used have two regular electrodes and copper wires. The wires extend beyond the end of the sock to connect to the data acquisition system*

The sock was tested using simulated walking, this was used as opposed to actual walking because of limitations in the measuring set-up and to prevent the thin copper wires from breaking. When measuring two sensors at the same time the signals interfere with each other. There are multiple aspects that can cause this interference. First of all, inside the DAQ there is an analog-to-digital converter(ADC), this converter causes a settling time in the signal and "stores" previous inputs when saving data. When the sampling frequency is high, the signals interfere too much with each other. The measurement frequency was set to 10Hz, this resulted in the output in Figure 15. It also became clear that noise from the environment influences the signal, the noise was reduced by intertwining the wires.

Figure 15: *Output voltage of both heel and toe sensor of initial sock design during simulated walking.*

7.1.0.1 Conductive yarn

In order to improve the design the copper wires had to be replaced by more durable wires which would not cause discomfort. Conductive yarn was chosen as a possible solution, initially a Shieldex® conductive 2-ply yarn without coating was tested.

Due to the characteristic of the yarn to change resistance when stretched this yarn was not suitable. Another conductive yarn by Shieldex® which was coated by a TPU layer was also tested. This yarn(Shieldex® 117/17 x2 HCB TPU) showed improved results, where the coating prevents stretching and also protects the wires from possible short circuit due to touching wires. The wires have an increased resistance compared to copper wires, the resistance measured is approximately 300 ohm/m. Compared to the resistance of the copper wires(5ohm/m) this is a significant increase in resistance. It was determined based on the length of the wires needed, which is at most 50cm per sensor resulting in a resistance in the yarn of 150ohm, that the increase in resistance is sufficiently small for this use-case.

7.1.0.2 Improved design

Based on the initial design an improved design for the sock was made. It was decided to use conductive yarn until the heel of the sock, and to make a connection at the heel of the sock to attach the data acquisition device. Attaching and removing the data acquisition device to the sock should be easy and yet the connections between the wires on the sock and the device need to be solid. It was decided to use conductive buttons as a connection method, see Figure 16 Figure 17. Furthermore, based on paragraph 7.0.3.1 it was decided to use sensors with interdigitated electrodes. In Appendix F two additional pictures of the final sock design are available.

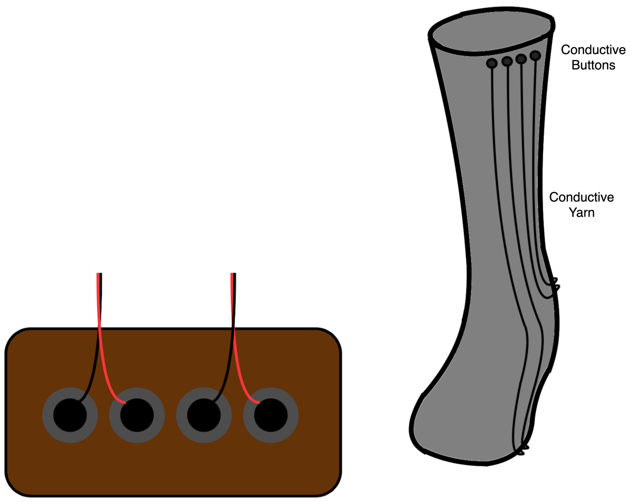


Figure 16: *Sock design including attachment system for data acquisition. This connection system using buttons could be suitable for a final design where the data acquisition device is attached to the ankle.*

During final testing of the sensorized sock it became clear that the electrodes(Both IDE and normal) cause some issues when walking. The pressure distribution on the sensor and electrodes can influence if the signal is positive or negative. Furthermore, by design of the sensor, where the pressure of the encapsulation ensures the wires and electrodes make a connection, the connections are prone to damage due to the weight and friction exerted on the sensor due to walking. A few tests were performed on the final sensor design, after which unfortunately the heel sensor became defective. Due to time limitations, it was not possible to improve the design to ensure better durability of the sensors during walking.

Figure 17: *Picture of final fabricated sock with buttons as connectors and padding on the sensors*

7.2 Algorithm development

In Matlab an algorithm is developed to determine when steps occur, and based on the data the gait parameters of the person walking are calculated. The Matlab algorithm to obtain the data is available in Appendix D and the algorithm calculating the gait parameters is available in Appendix E. The gait cycle exists of multiple phases as can be seen in Figure 18, in this diagram it becomes clear how each heel strike and toe off determines where we are in the cycle.

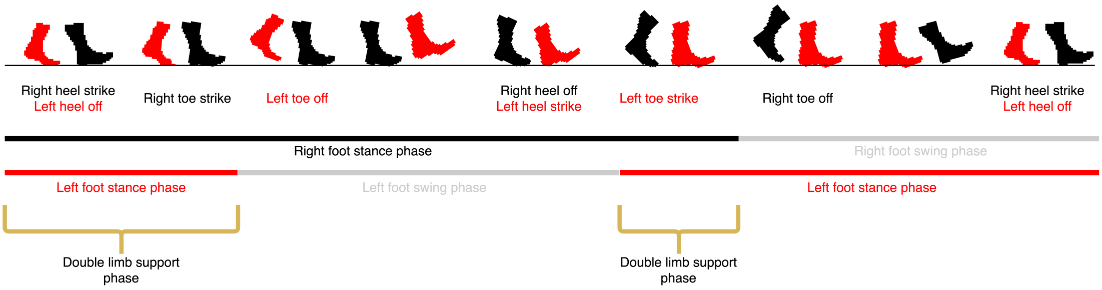


Figure 18: *Gait cycle with gait detection events, namely: Left heel strike, Left toe off, Right heel strike and Right toe off, based on these events the gait parameters are calculated. The left foot is displayed in red and the right foot is displayed in black.*

7.2.1 Input parameters

The data used as input into the algorithm is: Time, Left foot heel data, Left foot toe data, Right foot heel data, and Right foot toe data. Based on these inputs the occurrence of heel strikes/offs and toe strikes/offs of each foot can be determined. In the following section, an explanation is given regarding how heel strikes are registered based on the voltage output of the heel sensor.

7.2.2 Registering heel strikes

In Figure 19 a typical output of a sensor placed in the heel of the sock is displayed. A sharp incline in the graph indicates a heel strike, while a sharp decline indicates a heel off. In Matlab based on this principle an if statement is placed to determine whether a heel strike occurs by finding the slope of the graph. The following section of code adds a dummy heel strike to the vector LFH_dummy_strikes.

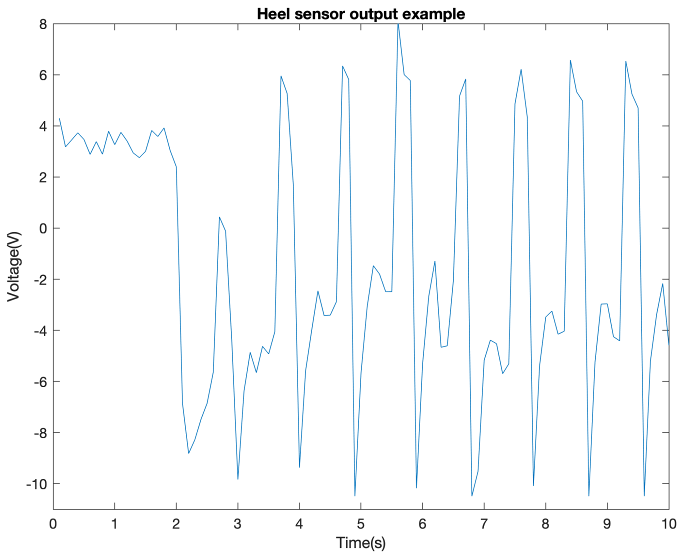


Figure 19: *This figure shows the voltage output of a heel sensor during simulated walking, a sharp incline indicates a heel strike while a sharp decline indicates a heel off*

```
if (LFH(k+1) -LFH(k)) > a
    LFH_dummy_strikes(end+1) = k;
```

Than the next section determines whether a dummy heel strike is a real heel strike based on the time between the previous registered heel strike and the current time.

```
if sum(ismember(k-V, LFH_dummy_strikes)) == 0
        LFHstrike = 1; % indicates a new step is
           possible
else      ,LFHstrike = 0; % indicates a step is too
   close to the previous one so not allowed
end
```

This final section adds a heel strike to the vector LFH_strikes if a toe off has passed since the last heel strike and if enough time has passed since the previous heel strike.

```
if LFHstrike == 1 && length(LFH_strikes) == length(
   LFH_offs)
```

```
            LFH_strikes(end+1) = k/10; % Adds a
               heel_strike to the vector heel_strikes if
               a heel_off has taken place and the
               distance to the previous heel strike is
               less than a*0.1
    end
end
```

Similar to the previous code all other strikes and offs of the heel and the toe of both feet are determined.

7.2.3 Determining starts and ends of individual steps

A step is counted from heel strike until toe off. This is the period during which the foot touches the ground. The start and end timestamps of each step for both the left and right foot are registered in vectors. Some conditions are set for a consecutive heel strike and toe off to register as a step, since if both events are too far apart they are not useful for analysis. So, if the time difference between heel strike and toe off is less than 0.8 seconds a step is counted. Otherwise, the step is omitted.

7.2.4 Cadence

The cadence is calculated with two different methods. One method makes use of the number of steps counted in the last five seconds and updates the average of this period as the cadence at the end of this period. Another method used is the fast Fourier transform, which is used to efficiently calculate the discrete Fourier transform(Dirican and Aksoy, 2017). In this method, the signal is converted from the time domain to the frequency domain. Where the dominant frequencies can be displayed as peaks. Since gait has a relatively small range of frequencies that occur around 1 step/s, the dominant frequency within a range around this frequency is taken as the walking frequency. This number is converted to steps per minute.

7.2.5 Swing/stance time

The swing time is calculated based on the amount of time a foot is in the air, this is the time difference between toe-off and heel-strike of one foot. The stance time is calculated based on the time a foot touches the ground. this is the time difference between heel-strike and toe-off. The swing and stance time are only taken into account if the time difference between two steps of a foot is less than 2.5 seconds, this ensures that for pauses during walking no gait cycle parameters are being calculated. At the end of the algorithm, the mean of all swing times is taken to calculate the average swing time.

7.2.6 Double limb support time/phase

The double limb support time can be calculated for each foot. For the left foot, the double limb support time is the time during which apart from the left foot the right foot also touches the ground. The double limb support time for the left foot is taken as the sum of Stepsend-right(i) - steps-left(i+1) and Stepsend-left(i+1) - steps-right(i+1). To calculate the double limb support phase the double limb support time is divided by the length of the gait cycle of the left foot. Then the mean of all the double limb support phases is taken to calculate the average double limb support phase.

7.2.7 Stride time variability

Stride time variability is the variance of each stride time. This is calculated by taking the mean of all stride times, then calculating the standard deviation of each stride time compared to the mean. And finally the stride variability is calculated by dividing the standard deviation by the mean stride time and multiplying by 100.

7.2.8 Output of the algorithm

The final output of the algorithm are statements with the most important calculated gait parameters and the indicators for whether a person does or does not have Ataxia. In Figure 20 the outputs are shown, it can be seen that the initial indication is that the person does not have ataxia. This is based on Cadence, swing-phase, and stride time variability. If two or more of these indicators are deemed ataxic the initial indication is also that the person has Ataxia, if one or fewer of the indicators indicates Ataxia the person is deemed as healthy or having no Ataxia. In this example experiment, the average stance phase was determined to be 55%, the swing phase was 45% the double limb support phase was 25%, and the average cadence was determined to be approximately 42 steps per minute.
Furthermore, in Figure 21 the voltage output of the sensors is displayed. Furthermore, the red, green, blue and black triangles indicate respectively the left heel strike, left toe off, right heel strike and right toe off. It can also be seen that based on these indicators the steps of the left and right foot are displayed as black and red lines respectively. Additionally, the purple line displays the cadence.

```
The average stance phase is 55.0336%
The average swing phase is 44.9664%
The average double limb support phase is 24.6324%
The average cadence over the whole experiment is 42.2857 steps per minute
Based on the cadence, gait cycle and stride variability, the initial indication is you have no Ataxia
Your cadence is lower than expected for a healthy person
Your swinphase is conforming a healthy persons gait cycle
The variability in your stride phase is conforming a healthy persons gait
```

Figure 20: *The text outputs of the algorithm in Matlab give the most important gait parameters and give an Ataxia indication based on set thresholds.*

8 Discussion & Conclusion

8.1 Sensor fabrication and testing

Thin piezoelectric sensors based on a PVDF-TPU substrate were developed, during the research multiple iterations were used to optimize the design of the sensors.

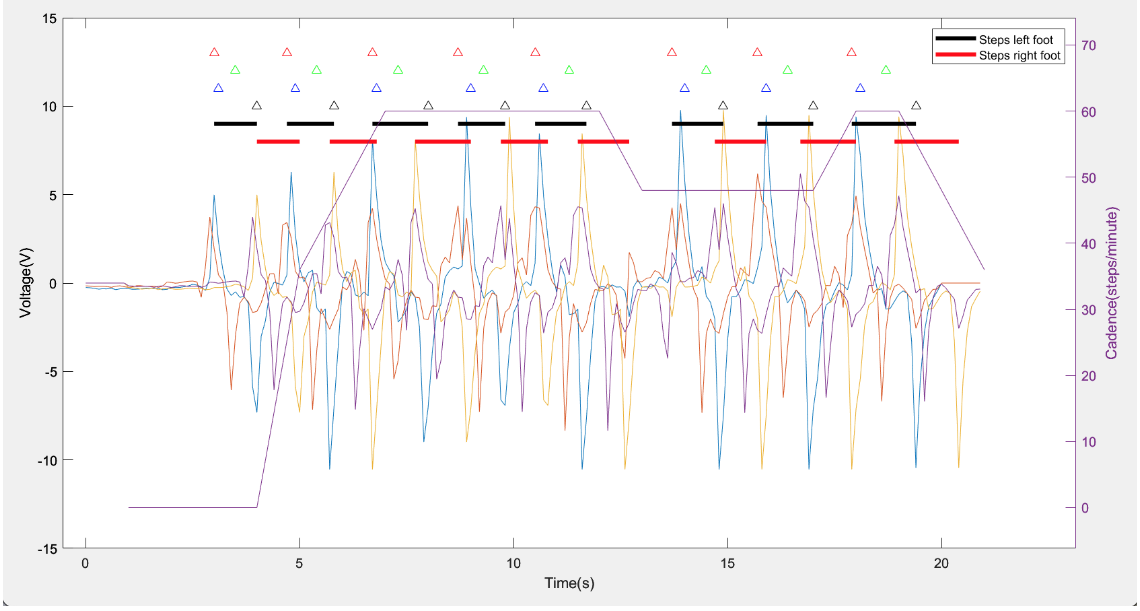


Figure 21: *The graph outputted by the algorithm shows each previous step recorded and displays the cadence over the time of the experiment. The triangles display each heel strike and toe off for the left foot, Left-foot-heel-strike(red), Left-foot-heel-off(green), Left-foot-toe-strike(blue), Left-foot-toe-off(Black)*

Sensors with a substrate of a higher thickness showed a more linear output, while the sensitivity compared to sensors with thinner substrates remained similar. The output for sensors with a high thickness was linear for at least the range from 60N to 180N. Interdigitated electrodes were introduced to increase the voltage output of the sensors, in Figure 10 it was seen that these electrodes have a higher sensitivity than sensors with normal electrodes. The sensitivity of IDE sensors was approximately 0,052 Volts/Newton, while the sensitivity of the sensors with normal electrodes was measured to be 0,031 Volts/Newton, this means the sensitivity of the interdigitated electrode sensors was 1,7 times higher compared to normal electrodes. The reliability of the sensors was determined by applying 1000 cycles of a 100N load. Sensors with normal electrodes showed good reliability after 1000 cycles, minimal drift occurred during the experiments. The sensors with interdigitated electrodes showed slightly less reliability, the sensors showed output during compression for 1000 cycles, although the magnitude of the voltage output was varying more compared to sensors with normal electrodes. This was caused by the pressure acting on the interdigitated electrodes, which did not occur on the normal electrodes.

8.2 Sensorized sock

The sensors were integrated into a regular sock, conductive yarn(Shieldex® 117/17 x2 HCB TPU) was used as wires and a button system was designed as a method of connecting the sock to a data acquisition system. The designed prototypes showed voltage spikes corresponding to the steps taken. It became apparent during testing of the prototypes that the sensors are prone to breaking under the force/friction generated by walking. Especially the electrodes were a cause of interference during walking, the connection was not very solid due to the fabrication method and pressure on the electrodes caused further issues with determining the correct events. Furthermore, even though in the scope of the research the data acquisition module for measuring the voltage output was not taken into account, during experiments it became clear that it is and will be a challenge to create a reliable compact method of measuring the output of piezo-electric sensors. Padding was created underneath the sensors to reduce the impact on the sensors, and especially prevent the force of the wire pressing on the electrode to cause for issues. The output was clear for a few experiments, after which further breakage occurred.

8.3 Developed algorithm

An algorithm was developed to convert the obtained voltage output into gait parameters. Two algorithms were created, one for live measuring and display of voltage output and one for converting the obtained data in gait parameters. The stride time, swing time, stance phase, swing phase, double limb support phase, cadence, and variability were obtained by the algorithm. The algorithm was designed in such a way that based on the gait parameters obtained an indication could be given whether a patient does or does not have Ataxia. For this indication the gait parameters cadence, swing phase, and variability were used. If two or more of these indicators indicate Ataxia, the patient is given the indication it might have Ataxia. The algorithm also generates a plot where the voltage output for each sensor is displayed and where it can be seen how this output converts to toe-strikes, heel-strikes, toe-offs and heel-offs for each foot. This plot also displays the average cadence during the experiment.

8.4 Feasibility and future research

Although the prototype was able to detect gait parameters it was not durable and issues are present with measuring piezoelectric output. The sensorized sock developed in this research can be compared with other similar products that it would compete with subsubsection 4.1.5. Taking into account the challenges that arise by having to develop very thin sensors that are durable as well, it becomes clear that making insoles is cheaper, more durable and gives more design options in choosing types of sensors(i.e. 3D/thicker sensors can be used, for example capacitive sensors). The advantage of the sensorized sock compared to the insoles is the fact you can measure gait without the patient wearing shoes, this advantage might not outweigh the benefits insoles have over socks. Especially, when taking into account that a data acquisition system also has to be attached to the sock, which when using insoles could be integrated into the shoe or in the insoles. Future research could be done into developing sensorized socks based on piezoelectric sensors for measuring gait when the data acquisition limitation occurring during measurement of piezoelectric sensors are solved. Taking all factors into account future research should be more focused on insoles currently, until further advancements are made into thin flexible sensors.

References

Amitrano, F., Macagno, M., Rossotti, S., Viganò, D., Cesarelli, M., and D'Addio, G. (2022). E-textile smart socks for gait analysis: a preliminary validation study. *Gait & Posture*, 97:26.

Ataxia.org (2022). What is ataxia?

Buckley, E., Mazzà, C., and McNeill, A. (2018). A systematic review of the gait characteristics associated with cerebellar ataxia. *Gait & posture*, 60:154–163.

Chen, J.-L., Dai, Y.-N., Grimaldi, N. S., Lin, J.-J., Hu, B.-Y., Wu, Y.-F., and Gao, S. (2022). Plantar pressure-based insole gait monitoring techniques for diseases monitoring and analysis: A review. *Advanced Materials Technologies*, 7(1):2100566.

Dirican, A. C. and Aksoy, S. (2017). Step counting using smartphone accelerometer and fast fourier transform. *Sigma J. Eng. Nat. Sci*, 8:175–182.

Dong, T., Guo, Y., Gu, Y., Wang, L., Liu, T., Wang, X., Li, P., and Li, X. (2021). Design of a wireless and fully flexible insole using a highly sensitive pressure sensor for gait event detection. *Measurement Science and Technology*, 32(10):105109.

Ge, D., Babangida, A. A., Hu, Z., Zhang, L., and Wang, M. (2021). Flexible pressure sensor based on a thermally induced wrinkled graphene sandwich structure. *IEEE Sensors Journal*, 22(4):3040–3051.

Grobe-Einsler, M., Taheri Amin, A., Faber, J., Schaprian, T., Jacobi, H., Schmitz-Hübsch, T., Diallo, A., Tezenas du Montcel, S., and Klockgether, T. (2021). Development of sarahome, a new video-based tool for the assessment of ataxia at home. *Movement Disorders*, 36(5):1242–1246.

Hartono, A., Darwin, Ramli, Satira, S., Djamal, M., and Herman (2016). Electric field poling 2g v/m to improve piezoelectricity of pvdf thin film. In *AIP Conference Proceedings*, volume 1719, page 030021. AIP Publishing LLC.

Jaber, M. A. and Massicotte, D. (2009). A new fft concept for efficient vlsi implementation: Part ii-parallel pipelined processing. In *2009 16th International Conference on Digital Signal Processing*, pages 1–5. IEEE.

Kadirvelu, B., Gavriel, C., Nageshwaran, S., Chan, J. P. K., Nethisinghe, S., Athanasopoulos, S., Ricotti, V., Voit, T., Giunti, P., Festenstein, R., et al. (2023). A wearable motion capture suit and machine learning predict disease progression in friedreich's ataxia. *Nature Medicine*, pages 1–9.

Lehmann, D. R. and Winer, R. S. (2008). *Analysis for marketing planning.*

Li, L., Zhang, M., Rong, M., and Ruan, W. (2014). Studies on the transformation process of pvdf from α to β phase by stretching. *Rsc Advances*, 4(8):3938–3943.

Lim, J. Y., Kim, S., and Seo, Y. (2015). Enhancement of β-phase in pvdf by electrospinning. In *AIP Conference Proceedings*, volume 1664, page 070006. AIP Publishing LLC.

Lincoln, L. S., Bamberg, S. J. M., Parsons, E., Salisbury, C., and Wheeler, J. (2012). An elastomeric insole for 3-axis ground reaction force measurement. In *2012 4th IEEE RAS & EMBS International Conference on Biomedical Robotics and Biomechatronics (BioRob)*, pages 1512–1517. IEEE.

Lucangeli, L., D'Angelantonio, E., Camomilla, V., and Pallotti, A. (2022). Sistine 2.0: Sensorized socks for postural telemonitoring. In *2022 IEEE International Workshop on Metrology for Industry 4.0 & IoT (MetroInd4. 0&IoT)*, pages 283–288. IEEE.

Nie, B., Huang, R., Yao, T., Zhang, Y., Miao, Y., Liu, C., Liu, J., and Chen, X. (2019). Textile-based wireless pressure sensor array for human-interactive sensing. *Advanced Functional Materials*, 29(22):1808786.

Ometov, A., Shubina, V., Klus, L., Skibińska, J., Saafi, S., Pascacio, P., Flueratoru, L., Gaibor, D. Q., Chukhno, N., Chukhno, O., et al. (2021). A survey on wearable technology: History, state-of-the-art and current challenges. *Computer Networks*, 193:108074.

Pavone, P., Praticò, A. D., Pavone, V., Lubrano, R., Falsaperla, R., Rizzo, R., and Ruggieri, M. (2017). Ataxia in children: early recognition and clinical evaluation. *Italian Journal of Pediatrics*, 43(1):1–9.

Pawlak, R., Korzeniewska, E., Koneczny, C., Ha, B., et al. (2017). Properties of thin metal layers deposited on textile composites by using the pvd method for textronic applications. *Autex Research Journal*, 17(3):229–237.

Saidani, S., Haddad, R., Mezghani, N., and Bouallegue, R. (2018). A survey on smart shoe insole systems. In *2018 International Conference on Smart Communications and Networking (SmartNets)*, pages 1–6. IEEE.

Sappati, K. K. and Bhadra, S. (2018). Piezoelectric polymer and paper substrates: a review. *Sensors*, 18(11):3605.

Schmitz-Hübsch, T., Du Montcel, S. T., Baliko, L., Berciano, J., Boesch, S., Depondt, C., Giunti, P., Globas, C., Infante, J., Kang, J.-S., et al. (2006). Scale for the assessment and rating of ataxia: development of a new clinical scale. *Neurology*, 66(11):1717–1720.

Sencadas, V., Moreira, M. V., Lanceros-Méndez, S., Pouzada, A. S., and Gregório Filho, R. (2006). α-to β transformation on pvdf films obtained by uniaxial stretch. In *Materials science forum*, volume 514, pages 872–876. Trans Tech Publ.

Sengupta, D., Pei, Y., and Kottapalli, A. G. P. (2019). Ultralightweight and 3d squeezable graphene-polydimethylsiloxane composite foams as piezoresistive sensors. *ACS applied materials & interfaces*, 11(38):35201–35211.

sensoriafitness.com (2023). Sensoria core pair.

Shehata, N., Nair, R., Boualayan, R., Kandas, I., Masrani, A., Elnabawy, E., Omran, N., Gamal, M., and Hassanin, A. H. (2022). Stretchable nanofibers of polyvinylidenefluoride (pvdf)/thermoplastic polyurethane (tpu) nanocomposite to support piezoelectric response via mechanical elasticity. *Scientific Reports*, 12(1):8335.

Stolze, H., Klebe, S., Petersen, G., Raethjen, J., Wenzelburger, R., Witt, K., and Deuschl, G. (2002). Typical features of cerebellar ataxic gait. *Journal of Neurology, Neurosurgery & Psychiatry*, 73(3):310–312.

Tirosh, O., Begg, R., Passmore, E., and Knopp-Steinberg, N. (2013). Wearable textile sensor sock for gait analysis. In *2013 Seventh International Conference on Sensing Technology (ICST)*, pages 618–622. IEEE.

Wegener, C., Hunt, A. E., Vanwanseele, B., Burns, J., and Smith, R. M. (2011). Effect of children's shoes on gait: a systematic review and meta-analysis. *Journal of Foot and Ankle Research*, 4(1):1–13.

Wieringa, R. J. (2014). Design science methodology for information systems and software engineering.

Xia, Y., Gu, H., Xu, L., Chen, X. D., and Kirk, T. V. (2021). Extending porous silicone capacitive pressure sensor applications into athletic and physiological monitoring. *Sensors*, 21(4):1119.

Zorgautoriteit, N. (2022). Overview of the dutch healthcare authority. *Nederlandse Zorgautoriteit*.

Appendices

A Supplemental figures: Normal sensor 51um

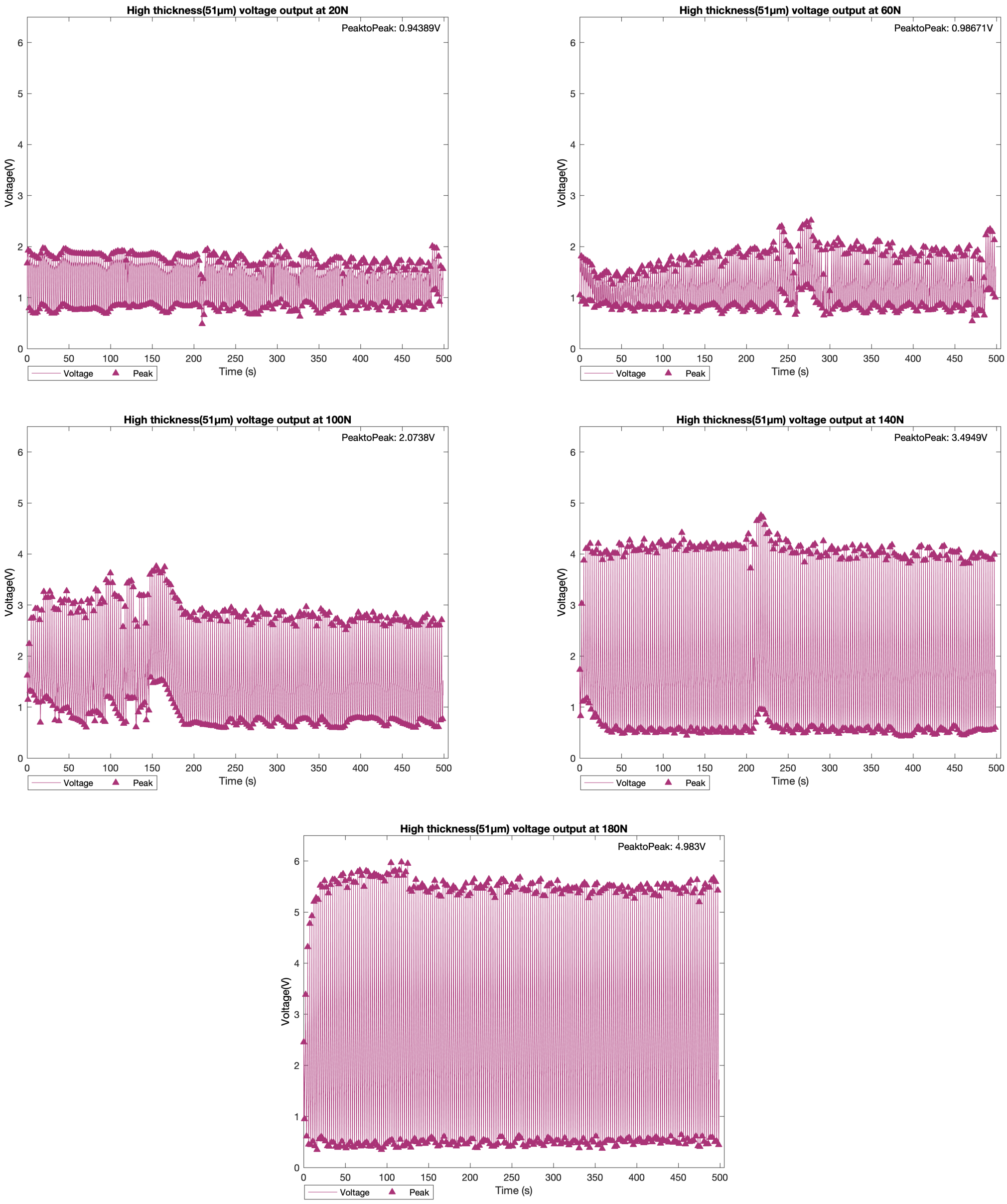


Figure 22: *Voltage output of a sensor with high thickness(51um), for different forces applied(20N-180N)*

B Supplemental figures: Normal sensor 34um

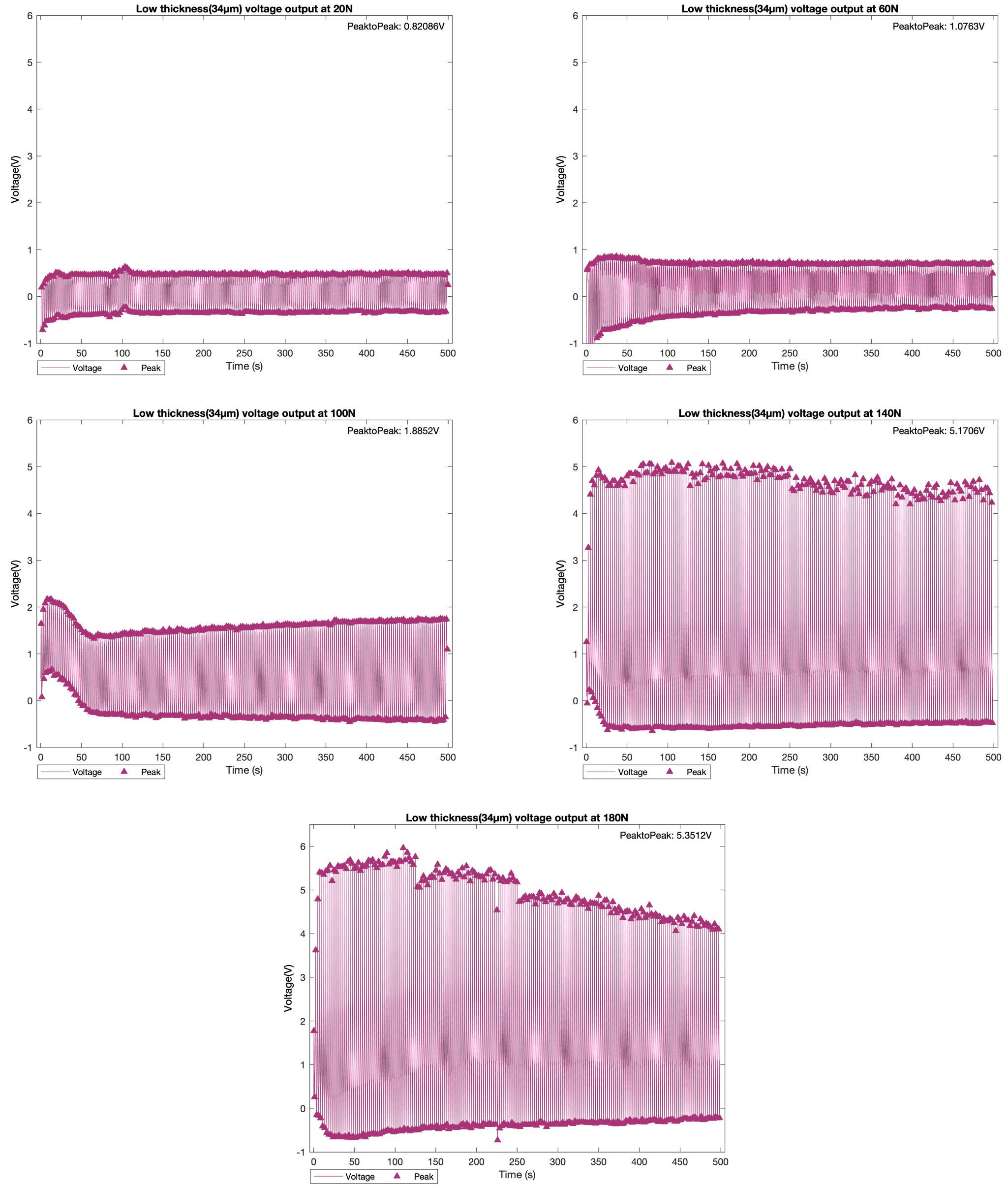


Figure 23: *Voltage output of a sensor with low thickness(34um), for different forces applied(20N-180N)*

C Supplemental figures: IDE sensor low thickness

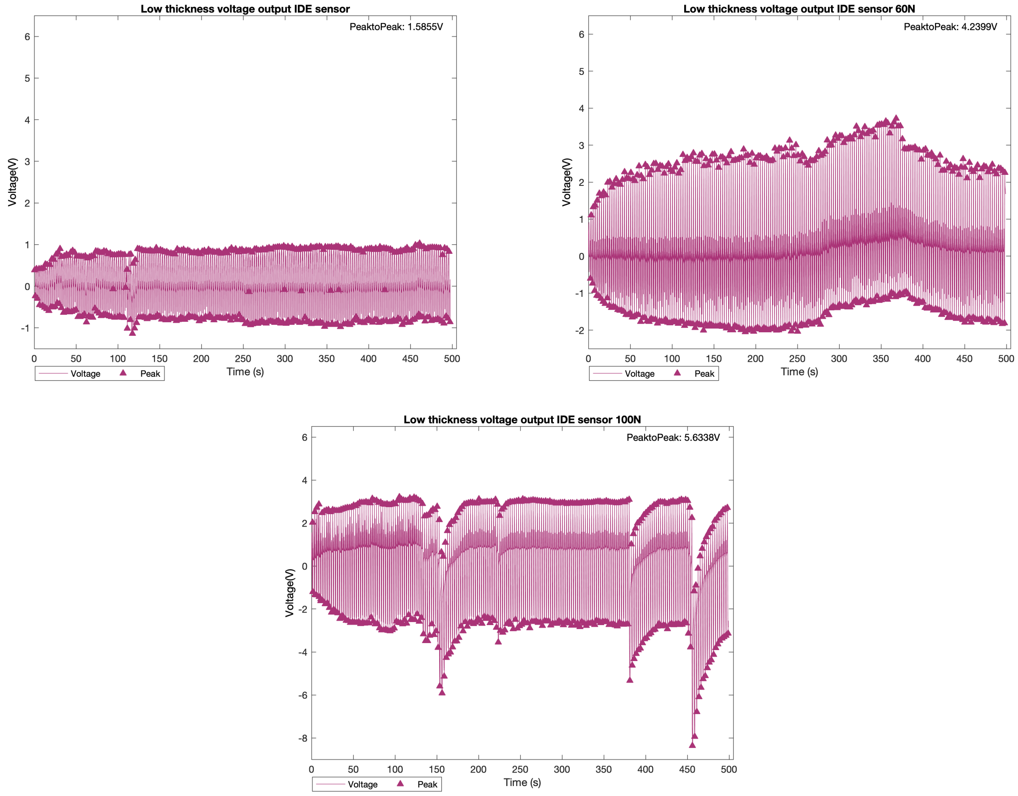


Figure 24: *Voltage output of a IDE sensor with low thickness, for different forces applied(20N-100N)*

D Algorithm Matlab code : Acquiring live data from sock

```
%% Algorithm for live view of voltage output and cadence
   of a person walking, cadence is based on one heel
   sensor %%

% This algorithm is only used for live viewing of the
   data, the obtained data
% can be inserted into another algorithm after
   acquisition in order to
% obtain all the gait parameters.

close all; clear all; clc

% create session and add analog input channels
```

```
d = daq(ni);
ch = addinput(d,Dev2,0,Voltage);
%ch = addinput(d,Dev2,1,Voltage);
ch = addinput(d,Dev2,2,Voltage);
%ch = addinput(d,Dev2,3,Voltage);

% set acquisition parameters
d.Rate = 10;

figure(2)
xlim([0, 20]);
ylim([0, 200]);

% create live data plot
figure(1);
ax = gca;
hold(ax, 'on');
plot1 = plot(ax, 0, 0);
plot2 = plot(ax, 0, 0);
xlabel(ax, 'Time (s)');
ylabel(ax, 'Voltage (V)');
xlim(ax, [0, 20]);
ylim(ax, [-10, 10]);

% define global variable to store acquired data
global data
data = [];

% define global variable to store cadence
global cadence
cadence = [];

% start acquisition and update plot
d.ScansAvailableFcn = @plotMyData;
start(d, Duration, seconds(20))

% define function to update plot
function plotMyData(obj, evt)
    % obj is the DataAcquisition object passed in. evt is
        not used.
    new_data = read(obj, obj.ScansAvailableFcnCount,
       OutputFormat, Matrix);
    global data
    data = [data; new_data];
    g = length(data);
    time = [0:0.1:(g/10 -0.1)];
```

```
    %time = read(obj, obj.ScansAvailableFCnCount,
       OutputFormat,Matrix);

    figure(1)
    plot(time, data(:,1), color, blue )
    plot(time, data(:,2), color, red)

    LFT = data(:,1); %toe
    LFH = data(:,2); %Left foot heel
L =20;
if length(LFH) > 20

        LFH2 = LFH(end-19:end);
        %time2 = time(end-19:end)

Y = fft(LFH2); % fast fourier transform

P2 = abs(Y/L);
P1 = P2(1:L/2+1); % one sided frequency plot
P1(2:end-1) = 2*P1(2:end-1);
P1(1) = 0;
a = (P1)==max(P1); % find strongest frequency
b = P1(a); % frequency
c = b*120; % cadence calculation from frequency
global cadence
cadence = [cadence;c];

else
    global cadence
    cadence = [cadence;0];
end
figure(2)
plot(time,cadence) % Live plot of time vs cadence
end
```

E Algorithm Matlab code : Computation and display of gait parameters

```
%% Sensorized sock algorithm
% This is an algorithm written by Jonah van der Meide in
   order to detect
% certain gait parameters with sensorized socks.
clc; close all; clear all;
```

```
deltax = 0; % Time to be removed from start of experiment
%% Data extraction from excel file(s)
Table1 = readtable('Testdataalgorithm.xlsx','Sheet','
   Sheet1'); % Sheet where the data can be found
Table1.Time = Table1.Time - deltax; % remove a period
   deltax of time from the start of the measurement
%Table1(Table1.Time < 0,:) = []; % remove all values for
   which time is negative
%Table1(Table1.Time > 40,:) = []; % remove all values for
    which time is above a certain value
x = Table1.Time; %Time
LFT = Table1.LFT; %Left foot toe
LFH = Table1.LFH; %Left foot heel
RFT = Table1.RFT; %Right foot toe
RFH = Table1.RFH; %Right foot heel

%% Set up arrays to store values detected from data
LFH_dummy_strikes = []; %Left foot heel strikes dummy
LFT_dummy_strikes = []; %Left foot toe strikes dummy
RFH_dummy_strikes = []; %Right foot heel strikes dummy
RFT_dummy_strikes = []; %Right foot toe strikes dummy

LFH_dummy_offs = []; %Left foot heel offs dummy
LFT_dummy_offs = []; %Left foot toe offs dummy
RFH_dummy_offs = []; %Right foot heel offs dummy
RFT_dummy_offs = []; %Right foot toe offs dummy

LFH_strikes = []; %Left foot heel strikes
LFT_strikes = []; %Left foot toe strikes
RFH_strikes = []; %Right foot heel strikes
RFT_strikes = []; %Right foot toe strikes

LFH_offs = []; %Left foot heel offs
LFT_offs = []; %Left foot toe offs
RFH_offs = []; %Right foot heel offs
RFT_offs = []; %Right foot toe offs

Steps_left = []; % Stores initial time values of left
   step
Steps_right = []; % Stores initial time values of right
   step
Stepsend_left = []; % Stores final time values of left
   step
Stepsend_right = []; % Stores final time values of right
   step
Steps = []; % Stores initial time values of all steps
```

```
Stepsend = []; % Stores final time values of all steps

Cadence = []; % Stores average cadence over the last five
    time periods

Swing_time_right = []; % Swing times of right foot
Swing_time_left = []; % Swing times of left foot
Stance_time_right = []; % Stance times of right foot
Stance_time_left = []; % Stance times of left foot
Double_limb_support_time = []; % Double limb support
   times for each step
Double_limb_support_phase = [];

Ataxiaindication = []; % Ataxia indication variable

%% Initial values determined by sensitivity of sensor etc
   ..

% Left foot toe
a = 2; % Positive slope or min val
b = 3; % Negative slope or min val

% Left foot heel
c = 2; % Positive slope or min val
d = 3; % Negative slope or min val

% Right foot toe
e = 2; % Positive slope or
f = 3; % Negative slope or

% Right foot heel
g = 2; % Positive slope or
h = 3; % Negative slope or

l = 4; % Periods of time needed between steps

V = 1:l ; % Vector to store period between steps
W = 1:l-1 ; % Reduced vector to store smaller period
   between steps

%% %%Start of algorithm%%
for k = 1:length(x)-1



%% Registering left foot heel strikes
```

```
if LFH(k) > a
        LFH_dummy_strikes(end+1) = k;

    if sum(ismember(k-V, LFH_dummy_strikes)) == 0
              LFHstrike = 1; % indicates a new step is
                 possible
    else      ,LFHstrike = 0; % indicates a step is too
       close to the previous one so not allowed
    end
        if LFHstrike == 1 && length(LFH_strikes) ==
           length(LFH_offs)
                LFH_strikes(end+1) = k/10; % Adds a
                   heel_strike to the vector heel_strikes
                    if a heel_off has taken place and the
                    distance to the previous heel strike
                   is less than a*0.1
        end
end
```

```
%% Registering left foot heel offs
if LFH(k) < -b
        LFH_dummy_offs(end+1) = k;

    if sum(ismember(k -W, LFH_dummy_offs)) == 0
              LFHoff = 1; % indicates a new step is
                 possible
    else      ,LFHoff = 0; % indicates a step is too close
        to the previous one so not allowed
    end
        if LFHoff == 1 && length(LFH_strikes) == length(
           LFH_offs) + 1
                LFH_offs(end+1) = k/10;
        end
end
```

```
%% Registering left foot toe strikes
if LFT(k) > c
        LFT_dummy_strikes(end+1) = k;

    if sum(ismember(k-V, LFT_dummy_strikes)) == 0
              LFTstrike = 1; % indicates a new step is
                 possible
    else      ,LFTstrike = 0; % indicates a step is too
       close to the previous one so not allowed
    end
```

```
        if LFTstrike == 1 && length(LFT_strikes) ==
           length(LFT_offs)
                LFT_strikes(end+1) = k/10; % Adds a
                   toe_strike to the vector toe_strikes
                   if a toe_off has taken place and the
                   distance to the previous heel strike
                   is less than a*0.1
        end
end
```

```
%% Registering left foot toe offs
if LFT(k) < -d
      LFT_dummy_offs(end+1) = k;

    if sum(ismember(k -W, LFT_dummy_offs)) == 0
              LFToff = 1; % indicates a new step is
                 possible
    else      ,LFToff = 0; % indicates a step is too close
        to the previous one so not allowed
    end
        if LFToff == 1 && length(LFT_strikes) == length(
           LFT_offs) + 1
                LFT_offs(end+1) = k/10;
        end
end
```

```
%% Registering right foot heel strikes
if RFH(k) > e
        RFH_dummy_strikes(end+1) = k;

    if sum(ismember(k-V, RFH_dummy_strikes)) == 0
              RFHstrike = 1; % indicates a new step is
                 possible
    else      ,RFHstrike = 0; % indicates a step is too
       close to the previous one so not allowed
    end
        if RFHstrike == 1 && length(RFH_strikes) ==
           length(RFH_offs)
                RFH_strikes(end+1) = k/10; % Adds a
                   heel_strike to the vector heel_strikes
                    if a heel_off has taken place and the
                    distance to the previous heel strike
                   is less than a*0.1
        end
```

```
end

%% Registering right foot heel offs
if RFH(k) < -f
        RFH_dummy_offs(end+1) = k;

    if sum(ismember(k -W, RFH_dummy_offs)) == 0
              RFHoff = 1; % indicates a new step is
                 possible
    else      ,RFHoff = 0; % indicates a step is too close
        to the previous one so not allowed
    end
        if RFHoff == 1 && length(RFH_strikes) == length(
           RFH_offs) + 1
               RFH_offs(end+1) = k/10;
        end
end

%% Registering right foot toe strikes
if RFT(k) > g
        RFT_dummy_strikes(end+1) = k;

    if sum(ismember(k-V, RFT_dummy_strikes)) == 0
              RFTstrike = 1; % indicates a new step is
                 possible
    else      ,RFTstrike = 0; % indicates a step is too
       close to the previous one so not allowed
    end
        if RFTstrike == 1 && length(RFT_strikes) ==
           length(RFT_offs)
               RFT_strikes(end+1) = k/10; % Adds a
                  toe_strike to the vector toe_strikes
                  if a toe_off has taken place and the
                  distance to the previous heel strike
                  is less than a*0.1
        end
end


%% Registering right foot toe offs
if RFT(k) < -h
      RFT_dummy_offs(end+1) = k;

    if sum(ismember(k -W, RFT_dummy_offs)) == 0
              RFToff = 1; % indicates a new step is
```

```
                possible
    else     ,RFToff = 0; % indicates a step is too close
        to the previous one so not allowed
    end
        if RFToff == 1 && length(RFT_strikes) == length(
           RFT_offs) + 1
                RFT_offs(end+1) = k/10;
        end
end

end % end of algorithm
```

```
%% This section ensures that when the measurement ends
   with a heel/toe strike
% this final heel/toe strike is removed from the vector

if length(LFH_strikes) > length(LFH_offs)
        LFH_strikes(end) = [];
end

if length(LFT_strikes) > length(LFT_offs)
    LFT_strikes(end) = [];
end

if length(RFH_strikes) > length(RFH_offs)
    RFH_strikes(end) = [];
end

if length(RFT_strikes) > length(RFT_offs)
    RFT_strikes(end) = [];
end
```

```
%% Based on the strikes and offs the times of beginning
   and end of steps for each foot are calculated

% Left foot steps
for i = 1: length(LFH_strikes)
    for j = 1 : length(LFT_offs)
 if LFT_offs(j) - LFH_strikes(i) <= 1.5 && LFT_offs(j) -
    LFH_strikes(i) >= 0
        Steps_left(end+1) = LFH_strikes(i);
        Stepsend_left(end+1) = LFT_offs(j);
 end
    end
```

```
end
% Right foot steps
for i = 1: length(RFH_strikes)
    for j = 1 : length(RFT_offs)
 if RFT_offs(j) - RFH_strikes(i) <= 1.5 && RFT_offs(j) -
    RFH_strikes(i) >= 0
         Steps_right(end+1) = RFH_strikes(i);
         Stepsend_right(end+1) = RFT_offs(j);
 end
    end
end

% Combined steps
Steps = sort([Steps_left, Steps_right]);
Stepsend = sort([Stepsend_left, Stepsend_right]);


%% Cadence

for m = 5:length(x)/10

    index = find(Steps>m-5 & Steps <=m);
    Cadence(m) = numel(index)/5 * 60; % 120 because in
       this instance only one sock is measured
end

% mean cadence

Averagecadence = mean(Cadence);

%% Gait cycle parameters


% Left foot
for i = 1: length(Steps_left) - 1
 if Steps_left(i+1) - Stepsend_left(i) < 2.5
         Swing_time_left(end+1) = Steps_left(i+1) -
            Stepsend_left(i);
 end
end

for i = 1: length(Steps_left)
 if Stepsend_left(i) - Steps_left(i) < 2.5
         Stance_time_left(end+1) = Stepsend_left(i) -
            Steps_left(i);
 end
```

```
end


if length(Swing_time_left) > length(Stance_time_left)
        Swing_time_left(end) = [];
end

if length(Stance_time_left) > length(Swing_time_left)
        Stance_time_left(end) = [];
end

% Mean swing and stance phases for the left foot
Swingphaseleft = mean(Swing_time_left)/(mean(
   Stance_time_left) + mean(Swing_time_left)) *100;
Stancephaseleft = mean(Stance_time_left)/(mean(
   Stance_time_left) + mean(Swing_time_left)) *100;


% Right foot
for i = 1: length(Steps_right) - 1
 if Steps_right(i+1) - Stepsend_right(i) < 1
        Swing_time_right(end+1) = Steps_right(i+1) -
           Stepsend_right(i);
 end
end

for i = 1: length(Steps_right)
 if Stepsend_right(i) - Steps_right(i) < 1
        Stance_time_right(end+1) = Stepsend_right(i) -
           Steps_right(i);
 end
end

% Mean swing and stance phases for the right foot
Swingphaseright = mean(Swing_time_right)/(mean(
   Stance_time_right) + mean(Swing_time_right)) *100;
Stancephaseright = mean(Stance_time_right)/(mean(
   Stance_time_right) + mean(Swing_time_right)) *100;




% Double limb support time and phase
for i = 1: length(Steps_right) -1
 if Stepsend_right(i) - Stepsend_left(i+1) <= 2 &&
    Stepsend_right(i) - Stepsend_left(i+1) >= -2
```

```
        Double_limb_support_time(i) = (Stepsend_right(i)
           - Steps_left(i+1)) + (Stepsend_left(i+1)-
           Steps_right(i+1));
 else Double_limb_support_time(i) = 0;
 end
end



for i = 1: length(Double_limb_support_time) -1
    if Steps_left(i+1) - Steps_left(i) <= 2.5
        Double_limb_support_phase(i) =
           Double_limb_support_time(i+1)/(Steps_left(i+1)
            - Steps_left(i));
    else
        Double_limb_support_phase(i) = 0;
    end
end


MeanDBL = 100*mean(nonzeros(Double_limb_support_phase()))
   ;


% Stride variability

for i = 1: length(Steps)
     Stridetime(i) = Stepsend(i) - Steps(i);
end

Meanstridetime = mean(Stridetime);

for i = 1: length(Steps)
     STD(i) = (Stridetime(i) - Meanstridetime)^2    ;
end


Stride_variability = STD / Meanstridetime * 100;
%% Indication on gait being ataxic or not

% Cadence comparison                     ; Average steps/
   min below 105 (For a continuous walking test)
if Averagecadence < 105
    Cadenceindicator = 1;
else Cadenceindicator = 0;
end
```

```
if Cadenceindicator == 1
    Yourcadence = ' lower than expected for a healthy
       person';
else
    Yourcadence = ' sufficiently high for a healthy
       person';
end
% Swing phase comparison                 ; Swing phase
   below 38%


if Swingphaseleft < 38
    Swingphaseindicator = 1;
else Swingphaseindicator = 0;
end

if Swingphaseindicator == 1
    Yourswingphase = ' typical for a patient with ataxia'
       ;
else
    Yourswingphase = ' conforming a healthy persons gait
       cycle';
end


% Variability in stride time comparison  ; Variability
   above 5% Calculated
% by: Standard deviation devided by the mean

if Stride_variability > 5
    Stridevariabilityindicator = 1;
else Stridevariabilityindicator = 0;
end

if Stridevariabilityindicator == 1
    Yourvariability = ' higher than expected for a
       healthy persons gait';
else
    Yourvariability = ' conforming a healthy persons gait
       ';
end

if Stridevariabilityindicator + Swingphaseindicator +
   Cadenceindicator < 2
    Ataxiaindication = 'no Ataxia';
```

```matlab
else
    Ataxiaindication = 'Ataxia';
end



%% Print gait parameters and ataxia indication

D = ['The average stance phase is ', num2str(
   Stancephaseleft) , '%'];
E = ['The average swing phase is ', num2str(
   Swingphaseleft) , '%'];
F = ['The average double limb support phase is ', num2str
   (MeanDBL) , '%'];
G = ['The average cadence over the whole experiment is ',
    num2str(Averagecadence) , ' steps per minute'];

H = ['Based on the cadence, gait cycle and stride
   variability, the initial indication is you have ',
   num2str(Ataxiaindication)];

    I = ['Your cadence is', num2str(Yourcadence)];


    J = ['Your swinphase is', num2str(Yourswingphase)];


    K = ['The variability in your stride phase is',
       num2str(Yourvariability)];

% L = ['The maximum cadence measured is ', num2str(max(
   Cadence)), ' steps per minute'];
disp(D)
disp(E)
disp(F)
disp(G)
disp(H)
disp(I)
disp(J)
disp(K)



%% Print graph with information
```

```
plot(x,LFT)
hold on
plot(x,LFH)
plot(x,RFT)
plot(x,RFH)
xlabel('Time(s)')
ylabel('Voltage(V)')
plot(LFH_strikes, 13, '^r')
plot(LFH_offs, 12, '^g')
plot(LFT_strikes, 11, '^b')
plot(LFT_offs, 10, '^black')
% plot(Steps, 0, '*black')
for u = 1: length(Steps_left)
Leftsteps = plot([Steps_left(u) Stepsend_left(u)],[9 9],'
   color', 'black', 'Marker','none', 'LineStyle','-','
   LineWidth', 3);
end
for u = 1: length(Steps_right)
Rightsteps = plot([Steps_right(u) Stepsend_right(u)],[8
   8],'color', 'red', 'Marker','none', 'LineStyle','-','
   LineWidth', 3);
end
yyaxis right
ylabel('Cadence(steps/minute)')
plot(Cadence)

legend([Leftsteps Rightsteps],'Steps left foot','Steps
   right foot')
```

F Supplemental figures: Final sock attachment and sensor

Figure 25: *Picture of attachment on sock, a prototype of an attachment to connect the sock to a data acquisition system*

Figure 26: *Picture of a sensor with interdigitated electrodes and conductive yarn wires mounted on a sock*