Structural and Electrical Characterization of Dry Electrodes for Electrical Stimulation

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Dissertation presented to the School of Technology and Management of Bragança to obtain the Master's Degree in Industrial Engineering under the scope of the Double Degree agreement with UTFPR. This work was oriented by professor Dr. Paulo Jorge Pinto Leitão, from IPB, and professor Dr. Cristiano Marcos Agulhari, from UTFPR.

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"Talent is cheap; dedication is expensive. It will cost you your life."

- *Irving Stone, The Agony and the Ecstasy*

Dedication

This work is dedicated to my family. My parents Marcos and Selma, you always encouraged me to follow my dreams without fear. Andrea and Fabio, you were there every time i had to face the unknown and guided me.

To my grandmother Maria, who does not let me forget where I am from and still keeps inspiring me to challenge myself to greater heights. I hope you keep seeing me grow and evolve into the man you expect me to be.

Sara, you were always by my side, be it to comfort me or to prove me wrong. You stood by my side in moments I thought I was alone. You are greatly responsible for what I am today. You never stop inspiring me with your kindness, calm and patience. May we stay happy for as long as it takes.

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I thank my friends from Brazil specially for the time they spent listening to me and for the game nights. The friends I made in Portugal also have special thanks for their aid and attention. To Kathleen Carvalho, my most sincere gratitude for everything you made for me. Finally, my grandmother has special thanks for all her support even from afar.

Abstract

Wearables aimed at rehabilitation with signal monitoring and muscle stimulation support systems represent a great advancement toward an increase in the quality of health care. Common monitoring and stimulation usually use wet electrodes, but these have the problem of the gel dehydrating over time worsening the conductivity, then requiring the electrodes to be changed or verified frequently, increasing the workload of the professionals of the area. Other factors that increase the burden on them are the preparation, position, and connection of the electrodes since this usually requires specialized knowledge. To solve this issue, dry electrodes were developed and tested regarding endurance, signal acquisition quality, and electrical stimulation capabilities. The combinations of three substrates options, five different thin films, and three sizes were tested. Different connection methods were tested to enable the use of the electrodes with commercial standard connectors. Therefore, the goal of this work was to characterize the electrodes regarding signal acquisition quality, electrical stimulation, structural endurance, and connection stability. The electrodes were tested both in simulated environments as well as *in vivo* to better understand the influence of the body on them. The results show that with the current configuration, $TiNCu_{0.45}$ in Polylactici Acid (PLA), employing snap button connector is the best choice although still requires some structural improvements to keep the conductivity. Polyurethane (PU) and Cellulose were discarded due to their mechanical properties. The other thin films presented high susceptibility towards corrosion, which greatly degraded conductivity.

Keywords: Dry electrodes, sEMG, Functional Electrical Stimulation, Polymer

electrodes

Resumo

Vestíveis voltados à reabilitação com monitoramento de sinal e estimulação muscular representam um grande avanço em direção ao aumento da qualidade de saúde. Monitoramento e estimulação costumam usar eletrodos úmidos, mas estes têm o problema do gel desidratar com o tempo piorando a condutividade, exigindo que os eletrodos sejam trocados ou verificados com frequência, aumentando a carga de trabalho dos profissionais. Outros fatores que aumentam a carga sobre eles são a preparação, a posição e a conexão dos eletrodos, pois isso geralmente requer conhecimento especializado. Para resolver este problema, eletrodos secos foram desenvolvidos e testados quanto à resistência, qualidade de aquisição de sinal e capacidade de estimulação elétrica. Foram testadas as combinações de três opções de substratos, cinco filmes finos diferentes e três tamanhos. Diferentes métodos de conexão foram testados para possibilitar o uso dos eletrodos com conectores comerciais. Assim, o objetivo deste trabalho foi caracterizar os eletrodos quanto à qualidade de aquisição do sinal, estimulação elétrica, resistência estrutural e estabilidade da conexão. Os eletrodos foram testados tanto em ambientes simulados quanto *in vivo* para entender melhor a influência do corpo sobre eles. Os resultados mostram que com a configuração atual, $TiNCu_{0.45}$ em Ácido Polilático (PLA), empregando conector do tipo botão de pressão é a melhor escolha, embora ainda necessite de algumas melhorias estruturais para manter a condutividade. Poliuretano (PU) e Celulose foram descartados devido às suas propriedades mecânicas. Os demais filmes finos apresentaram alta suscetibilidade à corrosão, o que degradou bastante a condutividade.

Palavras-chave: Eletrodos secos, sEMG, FES (Estimulação Elétrica Funcional), Eletrodos de polímero

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Acronyms

Al2O3	Aluminum Oxide
Ca	Calcium
Cl	Chlorine
cm	Centimeters
Cu	Copper
EMG	Electromyography
sEMG	Surface Electromyography
FDM	Fusion Deposition Modeling
FES	Functional Electric Stimulation
g	Grams
K	Potassium
mm	Milimeters
Na	Sodium
PCB	Printed Circuit Board
PDMS	Polydimethylsiloxane
PEDOT	Poly(3,4 ethylenedioxythiophene)
PET	Polyethylene terephthalate
PLA	Polylactici Acid
PU	Polyurethane
PVDF	Polyvinylidene fluoride

pvCNT	Patterned Vertical Carbon Nanotube
rpm	Rotation Per Minute
SNR	Signal-to-noise-ratios
Ti	Titanium
$TiCu_{0.34}$	Titanium dopped with Copper
TiN	Titanium-Nitride
$TiNCu_{0.45}$	Titanium-Nitride dopped with Copper
WHO	World Health Organization

Chapter 1

Introduction

Technology inclusion on day to day life is each time more common and, nowadays, is considered something essential for quality of life. This reaches the point where depriving someone of technology even affects their participation and integration in society [1]. Older people are more susceptible to this modern problem since interaction with technology is often complicated and requires effort and dedication until you get used to each command. The development of easy-to-use devices has received attention however, fewer options for these devices were found concerning devices projected specially for elders. Even less was found if dealing with wearable technology due to its variety [2].

Among the most common devices, wearables with health purposes still have the aggravating of often requiring specialized knowledge for application and manipulation depending on the type of sensors and actuators implemented [3]. So wearables with unchangeable preset configurations represent an even bigger improvement in elders' lives.

For wearables with signal monitoring and stimulation capabilities, the electrodes' position have much influence on the output. Having its position fixed is unfeasible for disposable gelled electrodes (wet) as they need to be changed every session so dry electrodes emerge as an alternative.

1.1 Context and motivation

According to the World Health Organization (WHO), the number of people aged 80 years or older is expected to triple by 2050 reaching 426 million. Not only this but by 2020, it is estimated that there were more people aged over 60 than children younger than 5 years [4]. With the coming of age is not unusual to develop some kind of movement hindrance requiring other people's assistance.

The growth in the aging people numbers is not proportional to the growth of the number of professionals and clinics that can treat the deceases. In addition, many people injure their muscles in sports or accidents [5] increasing, even more, the number of patients. This may be seen as something trivial but the lack of access to good healthcare is a silent crisis that drastically affects the quality of life for these individuals [6]. What's more, the process of restoring functions, that is, rehabilitation, is a slow process, which may include different types of therapy [7], and, with the current easy access to information, today's consumer is demanding more options and higher quality healthcare. One promising treatment method is to inject bursts of short pulses to trigger action potentials [8]. Functional electric stimulation (FES) is one of the main types of electric stimulation but as expected it requires, again, specialized personnel and equipment.

To make the management of qualified people even more difficult, the disabilities in movement can come from multiple injuries or illnesses. Therefore, to understand the extent of the physiological impairments and the functional losses, one must first understand the nature of the interaction between the physical environment and the task being performed. To assist in visualizing muscle activation order and intensity electromyography (EMG) signals are acquired and interpreted by specialized staff, but this requires the patient to stay at the clinic for the duration of the exams. Besides the brief monitoring time, in hospitals or clinics, the patient is in a controlled ambient where there are few or no variables at all. This reduces the realistic indicators of the health of the patient [7] as some rare events are unlikely to occur. Therefore, a wider monitoring window is crucial to ensure that the desired events are recorded, e.g., a vital signal anomaly that occurs only when the person performs a specific given task. As it is unfeasible to maintain the patient at the clinical site and wear all the necessary sensors for long periods, an alternative that allows signal monitoring anywhere and at any time is extremely necessary. This is the demand for portable, versatile, and easy-to-use medical devices that can be used to do homecare, that is to say, wearables.

Implementing FES and EMG in wearables seems like an easy task as both technologies are already well developed but to do it in a way that allows the treatment session to be carried out at home by non-specialized people has proven to be a real challenge [9]. One of the biggest problems is the accuracy and endurance of the embedded sensors since the idea is for the patient to put on the wearable and clean it themselves. If disposable electrodes were used, when changing for new ones they may be misplaced or badly connected which can lead to problems with signal quality and stability. The impact they have on the skin is also something to pay attention to as electrolyte-dependent electrodes (also known as wet electrodes) may present abrasive characteristics as the electrolyte dehydrates over time. The use of conductive gel for long periods (more than sixty minutes according to the medical staff), beyond being uncomfortable may cause harm to the skin. Even if these risks did not exist, changes in conductivity may cause the worsening of signal acquisition and risk of "hot spots" [10], e.g. irregular distribution of current leading to discomfort or pain. On the other hand, dry electrodes are developed to properly work without the electrolyte, i.e., they do not present abrasive behavior even after hours of use.

This work is inserted into the NanoStim project, which the architecture can be seen in Figure 1.1. The block with “Sensor EMG” and “Actuator (FES)” represents the electrodes with these functionalities responsible for being the mediator between the patient’s body and the systems responsible for signal acquisition, electrical stimulation, and wireless communication. A more detailed description of the project’s architecture is presented by [11].

Therefore, this study describes the process of development and characterization of titanium-based electrodes and connectors for signal acquisition and electric stimulation capable of being embedded in wearable systems. The electrodes were implemented in

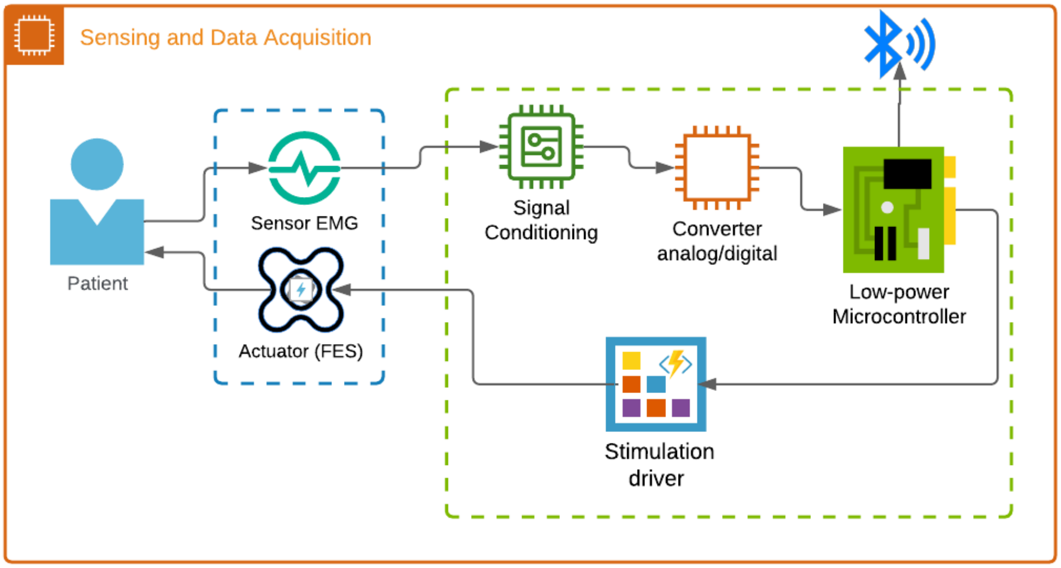

Figure 1.1: Architecture of the NanoStim solution

three different substrates, Polylactic Acid (PLA), Polyurethane (PU), and Cellulose. For PU and Cellulose only electrodes with 15 millimeters of diameter were developed while for PLA, electrodes with 34 and 7 millimeters of diameter were also implemented. Although they are referred to as titanium-based since the majority of them use titanium as an adhesive component for the thin film, an option with only copper (Cu) was implemented. The remaining thin film options all used titanium resulting in four more options, titanium (Ti), titanium-nitride (TiN), titanium dopped with copper ($TiCu_{0.34}$), and titanium-nitride dopped with copper ($TiNCu_{0.45}$).

1.2 Objectives

The objectives of this work are to develop connection methods that are robust enough to endure the whole process while maintaining the stability and signal quality. Another objective is to evaluate the proposed electrodes and integrate the most appropriate into the wearable.

The development of the technological solutions derived from these objectives is subject to the accomplishment of the following requirements:

1. Guarantee stable and reliable connection between electrodes and microelectronics
2. Ensure that signal-to-noise-ratios (SNR) are not degraded due to the mechanical connection
3. Ensure seamless integration of the electronic components into the wearable without compromising the comfort of the patient
4. Implement biocompatible thin films that keep low electrical resistance at the skin/electrode interface

1.3 Document structure

This work is divided into six chapters beginning with chapter 1, *Introduction*, where the context and motivation for this research and the objectives were described along with this structure detailing. Following, there is chapter 2, *Related Works*, approaching relevant advances in dry electrodes as well as giving a brief introduction and explanation about electric stimulation methods and EMG.

Chapter 3, *Development and experimental tests of connectors* introduce the necessity of connectors for the electrodes and the validation process followed for each designed option. In chapter 4, *Characterization methodology for electrodes*, the methodology followed to test each electrode and acquire the necessary data about their electrical and mechanical characteristics is accurately described. Chapter 5, *Analysis of experimental results*, brings the outcomes of the electrodes tests along with a brief explanation of what they represent and why they turned out that way. Chapter 6, *Conclusions and future work*, summarizes the whole project providing a conclusion and ideas for future works in the same research field.

Chapter 2

Fundamentals and references

This chapter aims to present novel and relevant works in the research fields explored by this work such as dry electrodes. Also, a brief definition of the concepts involved in the testing methodology and project is provided to assist in understanding the relevance and placement of this study.

2.1 Electromyography

Electromyography (EMG) is based on detecting and analyzing electrical signals stemming from skeletal muscles that, during muscle activation, present small electrical currents [12]. With the help of electrodes, these EMG signals known as electromyograms can be obtained.

EMG signal amplitude varies depending on the force exercised during the acquisition, i.e., to lift a small pebble would generate a much lower signal amplitude than lifting a 6 kg weight. This difference is resultant of the number of different muscles involved during the process since more motor units would be activated in proportion to the number of muscles [3].

There are various techniques for positioning the electrodes to acquire EMG signals from one specific muscle. One of the more common methods found in the literature was the bipolar configuration [13], which consists in placing two detecting surfaces (electrodes) between the innervation zone and the tendinous insertion. A third electrode is placed at an electrical neutral tissue, far from the EMG detecting ones, to serve as the ground for this signal [3], [14].

These electromyograms can be used as a diagnostic tool for neuromuscular diseases such as foot drop, low back pain and so on. Other than identifying the diseases, it can be used during rehabilitation, ergonomics research, and biofeedback [15] as an auxiliary parameter, but all of these require signal monitoring during the activities which can become a burden due to the amount of equipment required.

As technology advanced and quality of life increased, invasive electrodes for EMG acquisition started to get obsolete and unwanted since, in addition to harming the skin and muscle, they could cause discomfort, pain and stress during the insertion and movements [16]. With this in mind, methods to acquire the desired signals without damaging the body were created.

Placing specifically designed electrodes on the skin's surface right above the muscle and allowing the acquisition of electromyograms without invading the body is the main principle of surface electromyography (sEMG). However, as said, this process requires the use of electrodes different from the ones previously employed since it has much more electrical resistance between the electrode and the signal source than the invasive method.

The most common method to decrease the contact impedance between the electrode and the epidermis layer of the skin is to use electrolyte gels with the surface electrodes. The gold standard nowadays is Ag/AgCl electrodes (Figure 2.1) that already include the gel in the electrode's structure and are self-adhesive. These electrodes required little skin preparation and are easy to use even requiring specialized knowledge for electrode placement. These electrodes are disposable since they cannot withstand a long monitoring period as the signal quality decreases as the gel dehydrates increasing the impedance.

The stratum corneum is the upper layer of skin, composed mainly of dead skin, and is the major responsible for the increase in nonlinearity and resistivity. Using emery paper to lightly abrade the skin under the electrodes is sufficient to remove this layer and reduce dramatically the necessary voltage to stimulate the muscle or the amount of noise present

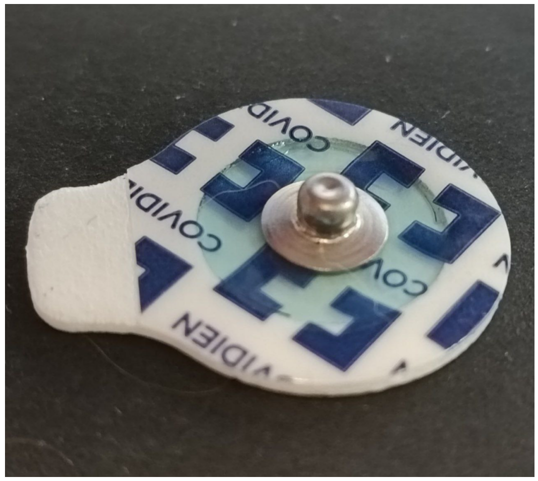

Figure 2.1: Commercial Ag/AgCl wet electrode

in acquired signals [17].

One decisive factor capable of greatly decreasing the treatment quality is the placement of electrodes. Positioning them in not abraded areas ends up in lesser results but positioning them in the wrong place can also lead to a phenomenon named crosstalk [3] in which the electrode acquires signals from more than one muscle at the same time. In worst cases, both electrodes can be wrongly placed resulting in completely disposable signals.

2.2 Functional electrical stimulation

Among the many types of electrical stimulation, FES has a bigger hole in later stages of rehabilitation. Differently from neuromuscular electrical stimulation (NMES) which is carried out in static positions with no functional movements [18], FES is usually applied to generate movements.

The loss of neural control over the muscles can be replaced to some degree by artificial stimulation [19]. Through a series of precisely timed bursts of electrical pulses, the electrical field triggers action potentials generating muscle contraction [8]. By cycling the FES application, higher exercise intensity can be achieved improving the results in cardiovascular conditioning and strength gain [20].

Not only for rehabilitation of muscles but FES can also be used to increase the circulation, maintain the range of motion, and facilitate the initiation of muscle activity [21]. For such things, implementing FES devices in wearables has become more and more common, especially in shirts and pants [22]. Such a thing is done by including electrodes rightfully placed above the desired muscles, but in some cases, to generate a functional movement, more than one muscle has to contract. To contract a group of muscles responsible for more complex movements, an array of electrodes may be used [23].

Using electricity to generate muscle contractions may cause discomfort and pain in the worst cases. To mitigate these sensations an array of constant currents sources can be used [24]. The current sources, in this case, are the electrodes used for stimulation that, if in higher numbers and with different sizes, reduce the sensation of artificial stimulation[23].

2.3 Dry electrodes

For both surface signal acquisition and electrical stimulation, the first electrodes to be employed were of the wet type, i.e., electrodes that require some type of conductive medium to achieve their purpose. This medium often comes as a gel to be easy to spread but, besides being uncomfortable, it may cause allergies and dehydrate over time worsening signal acquisition and increasing the risk of uneven current distribution during stimulation.

To dispense the use of the conductive medium, dry electrodes emerged as a non-invasive option. Although there are plenty of types and sizes of dry electrodes commercially available, the price and the difficult in integration for specific purpose systems made the researchers develop various new and improved types [25].

To ease the discernment between each type, an adaption of the categories division used for [26] will be used.

2.3.1 Textile electrodes

Another alternative to be used instead of wet electrodes is the so-called textile electrodes. These electrodes present cloth-like characteristics that allow them to be integrated into clothes or be used without arousing the awareness of other people [27].

Metal plated fabric

A common technique to make the common fabric into conductive fabric is to cover the fibers with conductive material. Three conductive fabrics based on polyester from Electro Magnetic Solutions Inc. were tested by [28]. The fabric was coated with a base combination of Polyethylene Terephthalate (PET), copper, and nickel but for one of the options, gold was added while for the other, carbon was used to enhance the conductivity and biocompatibility.

Ready-to-use commercial conductive fabric was also used by [29], but this one was acquired through [30]. Consisting of silver plated nylon, it allows the stretchability for both sides easing the placement and adjustment of the electrode.

Although called textile electrodes, they do not need to be strictly cloth, foam and materials similar to cloth also are included in this category. This is the case of [31], which tested five foams of three different materials, Polyethylene, Polyether, and a not specified one. Besides the material, each foam had its pores open or closed and the thickness altered to verify the adaptability to the skin surface. The conductibility was ensured by coating all surfaces of the foam with a 400 nm thick layer of silver. The thermal evaporation method used to accomplish this required an adhesion layer, that in this case was titanium.

Nanofiber web

Instead of covering normal cloth with conductive metals, [28] developed a nanofiber web electrode. It consisted of Polyvinylidene fluoride (PVDF)/$FeCl_3$ nanofibers with Poly(3,4 ethylenedioxythiophene) (PEDOT) polymerized over them. Another option was to polymerize silver instead of PEDOT but the metal plating process had to be done in specific ways to not use expensive or harmful chemicals. It resulted in a textile electrode with conductive properties intrinsic to its core instead of simply coated fibers.

Eeontex^{TM} was often found in the literature when searching for conductive textiles used in the manufacture of dry electrodes [32][33]. Most of their textiles are made out of nylon with a carbon cover. The differential feature is that the coating is made on a microfiber scale, so even if the electrical resistance is not as low as in other textiles, the endurance toward stretch and pressure makes it worthy of use.

2.3.2 Structured electrodes

Structured electrodes represent the type of electrode with intended protrusions. The goal may be to enhance the contact with non-penetrating electrodes or to reduce the skin impedance with penetrating electrodes [34]. Either way, the structures can be made by the electrode's main "body" material or by the material covering it such as metals or carbon.

Penetrating

Aiming to penetrate the *stratum corneum*, structures able to penetrate the first layer of the skin must have more than $10\mu m$ [35]. Structures resembling needles are the most common ones, being found with heights within $100\mu m$ [35] and $650\mu m$ [36] for easier penetration.

Square electrodes with 12mmx12mm were made by [37] with copper substrate covered by a thin layer of gold. The electrode was developed with an array of 10x10 microneedles, each with an average height of $320\mu m$.

Machining silicon wafer to obtain an array of 4x4 microneedles was the method used by [35]. After depositing titanium as the adhesion layer, a thin layer of iridium oxide was deposited to serve as the conductor and reduce the impedance at the skin/electrode interface. Figure 2.2 makes it easier to understand the dimensions of the microneedles.

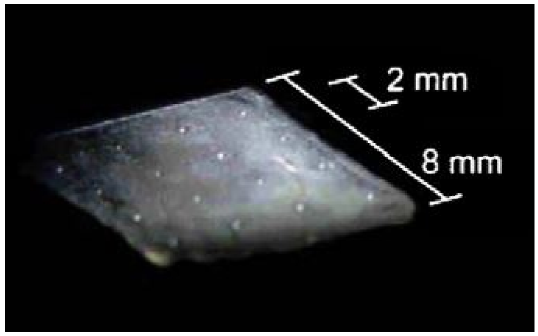

Figure 2.2: Micro needle electrode from [35].

Non-penetrating

The electrode implemented by [38] resembles the penetrating needles but with much greater height. What differentiates them is that [38] developed theirs for EEG acquisition as well. So, since it was supposed to go on the head, the amount of hair is greater as well. Needles with 3mm of height were 3D printed within an acrylic-based resin and, after the titanium adhesion layer, covered with a thin gold layer.

Although the author called it needles, the structure's features ended up enhancing the contact by overcoming the hair instead of penetrating the first skin layer. This is a result of the high number of needles making the electrode work with the same pressure principles of a bed of nails.

On the other hand, an electrode with a small number of needles was introduced by [38]. These needles had a flat point instead of a sharp one to just enhance the contact. Stainless steel was used as the main material for the electrode's body and the 12 needles on it, and a thin layer of silver was used to cover the electrode for its biocompatibility and conductivity. This electrode had some circuitry on its back, more specifically, an active filter which characterizes it as an active electrode.

With a high amount of structures, the Patterned Vertical Carbon Nanotube (pvCNT) electrode presented by [39] had a different appearance from the others since the structures were not as visible as the other ones. Figure 2.3 shows the electrode and allows the understanding of how its tubes differ from the previous needles. The image also shows how the electrode was fixed in a conductive tape instead of implementing some type of connector.

Figure 2.3: Patterned Vertical Carbon Nanotubes electrode fixed on conductive tape. Imagem from [39]

The carbon tubes were grown vertically in pillar formation after Al2O3 and iron were sputtered on the stainless steel foil substrate. The high amount of tubes with 1 to 1.5 mm in height allows some of them to break or deform without compromising the electrode capabilities. But this same amount can lead them to work as a solid surface invalidating the advantages of structured electrodes.

2.3.3 Film electrodes

Film dry electrodes possess the characteristic of being covered by extremely thin layers of conductive, biocompatible material such as gold, silver or titanium. The easy integration characteristic of film electrodes led them to be widely studied in recent years [40]. For being in direct contact with the *stratum corneum*, as time passes, the sweat on the skin's surface fills the gaps enhancing the contact, thereby reducing the noise [41].

For manufacturing the electrodes with film, there are many possible methods, including sputtering, deposition, electrostatic spinning, and chemical coating [26]. Usually, the film is made of conductive metals or carbon, but for any option, they need to be biocompatible to not cause skin abrasion.

Metal core

Surface electrodes often have metal and their derivatives as covers due to their excellent electrical conductivity. Besides the cover, the core of the electrode also can consist of metals such as copper. Due to the conductive capacity and low cost, copper would be an excellent material if it was not for its lack of oxidation resistance, but this problem is easily overcome by covering it with biocompatible materials.

A 100nm layer composed of 99.9% of pure gold is enough to protect the copper from oxidation while maintaining the low cost and conductivity quality. The wireless system developed by [42] and illustrated in Figure 2.4 employed this solution for EMG acquisition. By soldering three electrodes directly on the circuit board they ensured the stability of the connection but, as said by [25], it lacks wearability as the motion freedom was compromised.

Figure 2.4: Electrode composed of a copper core and gold thin film developed by [42].

Besides gold, silver is also widely used for medical purposes for being highly conductive, biocompatible, and a noble metal resistant to corrosion. Covering copper cores with it by electroplating was a technique used by [43] to implement an active electrode, i.e., an electrode with the circuitry integrated into it resulting in a modular, small device. [44] also used silver to cover the copper, but the core was plated with it instead. Also, instead of active, their electrodes were soldered into the acquisition systems and attached to a band-type holder made of elastic material.

Copper substrate covered with polypyrrole was used by [45], and although the core was made of copper, the cover made of conductive polymer brought some hindrances. The main problem with this combination was the poor surface connectivity and interference caused by hair [39].

Polymer core

Electrodes with polymer cores instead of metal cores usually present more varied shapes like the one developed by [46]. Using Polydimethylsiloxane (PDMS) as the precursor and Velcroat the ends to be easily wearable, this electrode was developed to be worn as a bracelet. Right in the middle of the PDMS, a convex shape protruding 1.5mm was implemented to enhance the contact of the metal pattern that would be deposited on top of it. Firstly, titanium was used as an adhesion layer, and then a gold pattern was implemented. This work used conductive glue to connect the wire with the metal pattern but the analysis did not distinguish between this type of connection and more usual ones.

Following the gold film idea, [47] used PDMS with gold on it as well. The main difference was the serpentine pattern used to allow the proposed electrode to be flexible even in the metal-coated parts. Snap button connectors were implemented in the PDMS directly in contact with the gold patterns to allow the use of commercial acquisition systems.

The electrode presented by [48] also takes advantage of the stretchability of PDMS by implementing serpentine patterned gold. It used copper above the gold to keep the conductibility without increasing the cost and to keep the adhesiveness. Figure 2.5 shows the electrode with the capability of being stretched, bent, folded, twisted, and compressed.

Besides gold film, other methods can be used to turn polymer substrates into electrodes such as silver nanowires [49]. The wires were inlaid below the PDMS surface to prevent them from delamination while maintaining the conductive and antibacterial properties of silver [50].

Among the techniques used to keep the polymer electrodes stretchable while using metal thin films (<100nm), some studies reported methods other than the former presented ones. These studies include meshing [51], surface roughing of the substrate [52], covalent bonding between metal and polymer [53], and foaming [54].

Apart from the three categories presented, there is still one that does not fit in either of

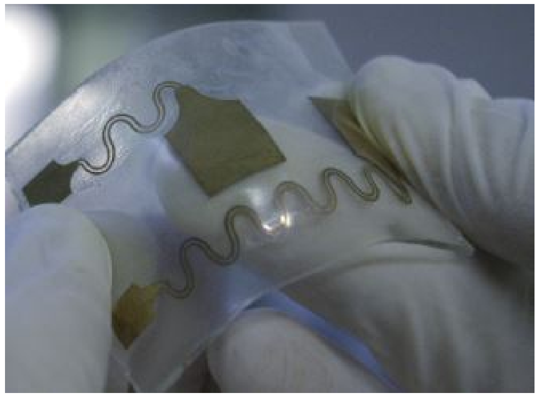

Figure 2.5: PDMS with gold serpentine patterned electrode. Example from [48]

them. That is because capacitive electrodes are considered dry electrodes but do not need to be in contact with the skin to acquire the signal. In return, they require complex active circuits for each sensor to be able to adapt to the constant changes in distance. The interfering noise from nearby equipment and the sensitivity necessary to detect the signal are also great hindrances when dealing with this kind of electrode.

The susceptibility of capacitive electrodes reaches the point of the acquired signal showing patterned artifacts referent to the breathing movement of the patient. This artifact usually is presented in low frequencies and can be removed both in analog and digital domains by applying a high-pass filter [55]. Other movements and the coupling of the patient's body with power lines also can cause electromagnetic interference. In compensation, these electrodes tend to be modular and have low-cost electronic interfaces [56] allowing the use of a flexible number of them.

Table 2.1 summarizes the most discerning electrodes studied in this state of art and allow a better comparison among them. The information about each electrode does not necessarily represent their limitations, it only shows for which purposes they were used so it is possible that some of them may be able to do more than what the table contains.

As noticeable in Table 2.1, the electrodes found in the literature have the most varied sizes. A study about the comfort while using two different electrode sizes was conducted by [57]. According to the volunteers, to generate the same muscle force output on the quadriceps and the hamstring, larger electrodes were more comfortable. Still, what makes

Table 2.1: Characteristics of the different dry electrodes found in the literature

Type	Main Material	Data Collection	Stimulation	Area (mm^2)
Coated polymer [48]	PDMS, gold and copper	EMG and ECG	NO	100
Coated polymer [46]	PDMS, gold and titanium	ECG	NO	1500
Coated polymer [49]	PDMS and silver	ECG and EMG	NO	400
Coated polymer [47]	PET and gold	EMG	NO	5
Capacitive [55]	Copper	ECG	NO	706,86
Core/shell [42]	Copper and gold	EMG	NO	6
Core/shell [44]	Copper and silver	EMG	NO	78.54
Core/shell [43]	Copper and silver	ECG	NO	490.87
Foam [31]	Conductive foam, silver and titanium	ECG	NO	314.16
Microneedle [37]	Copper and gold	EMG	NO	144
Microneedle [35]	Iridium-oxide	EEG	YES	64
Needle [38]	Resin, gold and titanium	EEG and ECG	NO	56.25
Needle [43]	Silver and stainless stell	EEG and ECG	NO	490.87
pvCNT [39]	Carbon, aluminum and iron	Impedimetric	YES	78.54
Textile [29]	Nylon and silver	-	YES	50
Textile [43]	Conductive fabric	ECG	NO	3375
Textile [28]	Copper and nickel	ECG	NO	64

the larger electrodes more comfortable to a certain point, and painful past that point was not conclusively analyzed [58].

Another noticeable fact is the number of electrodes for ECG acquisition. As long as the QRS complexes are discernible, the signal distortion may not be of importance since that, for ECG monitoring, the heart rate is the concern. With lower Signal-to-noise-ratios (SNRs) and the need to minimize noise, distortions, and artifacts, sEMG requires better preservation of the acquired signals [59].

Among the electrodes for sEMG acquisition, a large proportion is not used for electrical stimulation as these electrodes are applied in signal monitoring, not for rehabilitation, but for gesture identification [60][61][62]. The collected data can be used for both collaborative robotics and prosthesis, and for these, stimulation capabilities are not required.

Most of the literature found reports just the results for signal acquisition and/or electrical stimulation leaving outside the scope of the work the study about the influence of real situations. Sweat, continuous use for a long time, and the endurance of the electrodes are among the main subjects left outside the works as well as the connection method. Among the studies, only a few authors let it clear how the connection between the electrode and the circuitry was made, and even so, it was not in great detail.

As for the materials used, most electrodes have gold or silver in their composition. The ones that do not use these metals required some type of complex and, sometimes expensive process to enhance the conductibility and the biocompatibility. Even if the electrodes were not expensive, the fixation method often required unpractical and spacious equipment or needed to stay connected to other devices making it harder to monitor the patient outside clinical site.

Chapter 3

Development and experimental tests of connectors

This chapter aims to introduce the connectors development and improvement process along with the tests realized to validate them. The goal was to implement a connection method capable of enduring the treatment routine and cleaning processes without compromising the conductivity of the electrodes.

3.1 Alternatives for connectors

At the start of the project, three different connector alternatives were proposed. The first model was an electrode without a proper connection structure to be used with crocodile clips. As the tests were performed, the clips left dents on the thin film to the point of interfering with the conductivity. Figure 3.1 show one of the electrodes after some tests to exemplify the problem.

Another two types of connectors were drawn using *SolidWorks®*. Figure 3.2 shows the proposed connectors consisting in printed structures on top of the flat disk that composes the electrode. Figure 3.2.b was designed to be used with standard commercial snap button female connectors while Figure 3.2.a required a specific bracket to be placed on top of it and hold the wire in place. This bracket was also made through Fusion Deposition Modeling (FDM)

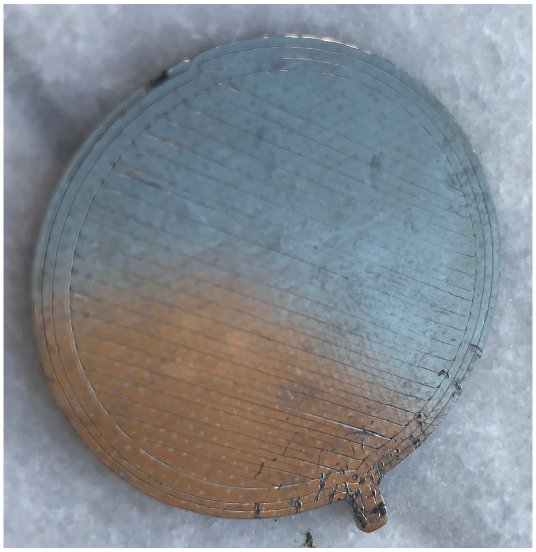

Figure 3.1: Damaged electrodes after multiple tests with crocodile clips.

with different sizes to better fit each electrode while maintaining the stability and easy-to-use characteristic. The 3D model and the way it should be used with the electrode can be seen in Figure 3.3.

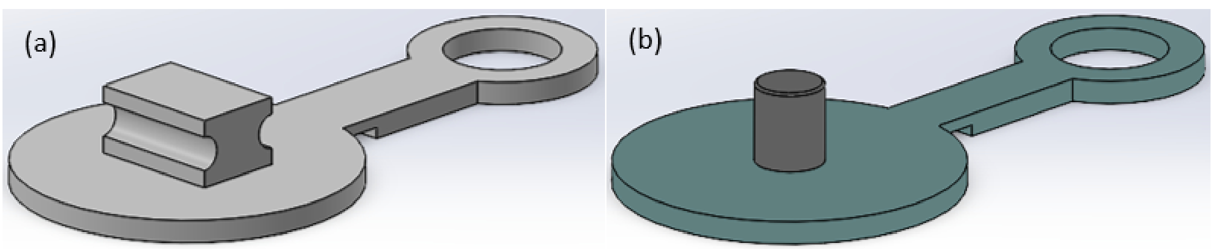

Figure 3.2: First connector options designed, (a) need the wire to be wound and (b) was supposed to be the male snap button connector

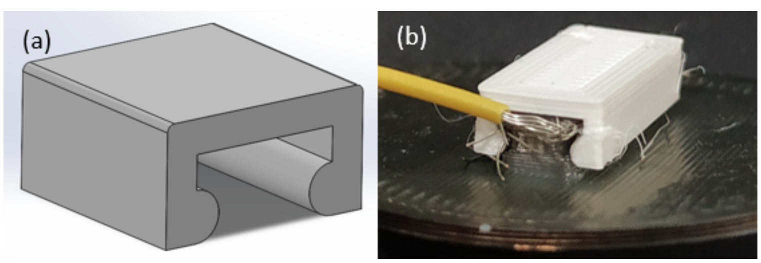

Figure 3.3: (a) presents the 3D model of the bracket and (b) show how it hold the wire in contact with the electrode connector

3.2 Longevity and stability experiments

Even though the crocodile clips damaged the thin films of the electrodes without connectors, they still ensured good conductivity and stability to some degree. The other connection methods were still prototypes, so some tests to validate them were necessary.

The first and simpler test was to connect and disconnect the electrodes to the cables. For the snap button, the female snap connector commercially available already had the cable integrated so the experiment consisted in simply attaching and detaching it. The connector with the bracket needed a wire to be manually wound at the electrode and then the bracket was attached to hold the wire in place as illustrated in Figure 3.3. Doing this repeatedly would allow to better verify the service life of the electrodes based only on connector integrity.

At the beginning of this test, the snap button developed proved to be unfruitful since it did not present any stability with the female connector and kept falling. Comparing the developed snap button with commercial ones made it clear that the shape chosen was too plain. Beyond not being spherical as the one from Figure 3.4.a, the irregularity of the surface shown in Figure 3.4.b compromised the conduction capabilities of the connector.

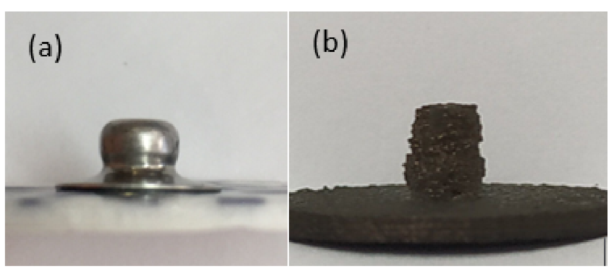

Figure 3.4: Comparison between (a) commercial male snap connector and (b) developed male snap connector

With the first snap button model, the connection endurance test should not present valid results as the pin was almost plain and kept falling off from the female connector. Even so, some cellulose electrodes with printing deformities on their connector applied a small pressure between the electrode and the female connector; sufficient pressure to break the connector with less than 20 cycles of docking and undocking. The connector did not properly break, but instead, the printing geometry used made it peel off of the electrode as shown in Figure 3.5.

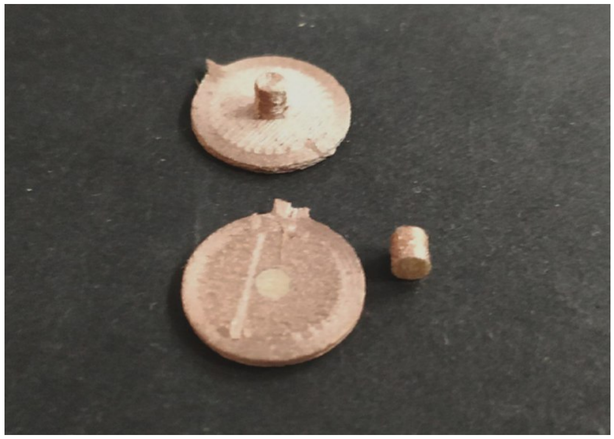

Figure 3.5: Cellulose electrode with its connector broken after several cycles of connecting and disconnecting

Since the pin break was caused by imperfections resulting from the printing process, cellulose electrodes were subjected to the following tests as well.

To solve the problem of stability a new version of the snap button was designed according to the internal dimensions of the female snap connector. The result, modeled as illustrated in Figure 3.6, was subjected to the same cycle of connecting and disconnecting several times.

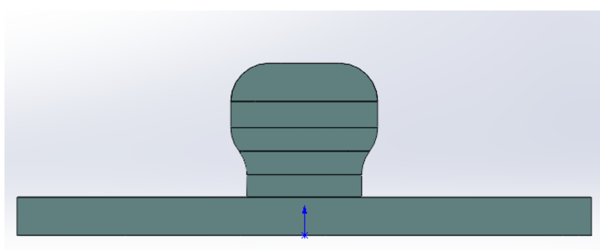

Figure 3.6: Snap button designed following the internal dimensions of the female snap connector

After making the adjustments on the snap button connector the tests were repeated and the new dimensions became a problem. Since it was designed with the exact internal dimensions of the female connector, the pressure exercised on the pin when inside the female snap was more than it could handle and the outcome showed in Figure 3.7 often occurred when trying to disconnect.

Even if the pin did not break, the pressure was enough to remove the thin film (see Figure 3.8) and, after a few disconnections, the thin film of the connector was heavily damaged.

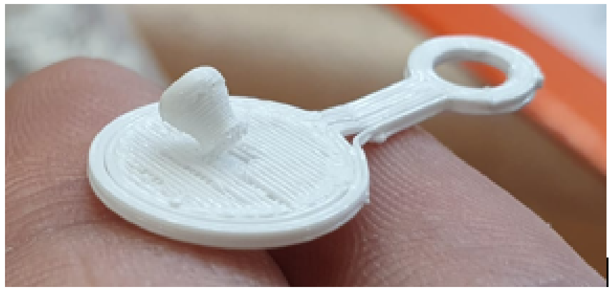

Figure 3.7: PLA snap button connector broken after disconnection

But even with this hindrance, the stability achieved the desired level, so the pin just had to have its diameter reduced. Other improvement implemented to ease the disconnection was the increase in the height of the pin.

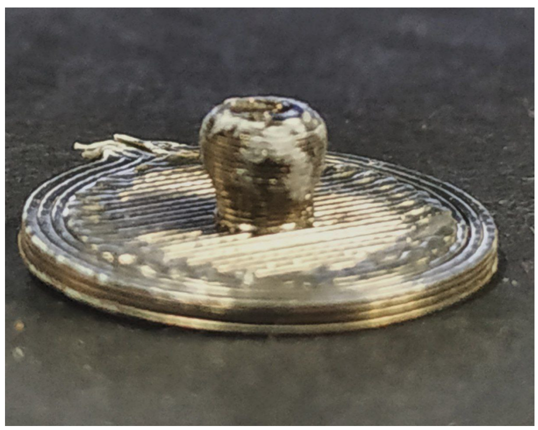

Figure 3.8: PLA snap button connector with damaged thin film

The connector with the bracket went through the same test but instead of the female connector, it had the wire wound around the salient part. The bracket was then fitted to fix the wire in place preventing it from falling off. For this test, both the wire and the bracket were removed and reconnected always trying to put them in the same spots.

Figure 3.9 presents a common result for the connector with bracket after the connection test. Although the electrodes were of different substrates, the brackets were made of PLA so, for PU electrodes the pressure made the connector deform and for cellulose, the bracket deformed. When the bracket was tested with PLA electrodes, for having the same properties, it did not deform and the pressure caused the bracket to break since it was the weakest structure among the two of them. During the tests, this type of connector also had the problem with the wire breaking as it folded for all sides at sharp angles.

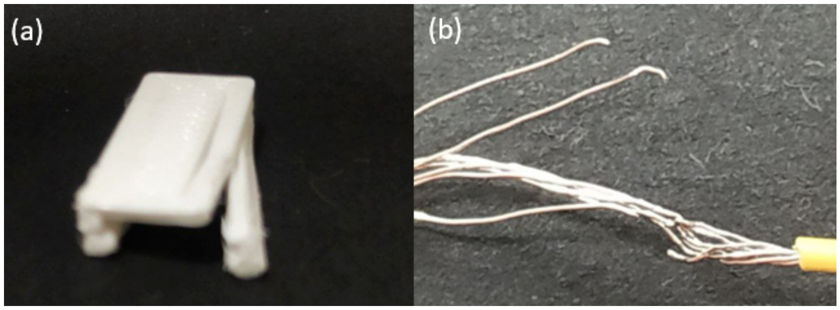

Figure 3.9: Broken bracket and wire after endurance tests

3.3 Conductivity tests

To analyze the conductivity and signal quality of each connector firstly a multimeter was used but the qualitative data was not enough to ensure which of them was safe to be used for electric stimulation. So, to obtain the characteristics of conductivity of the connectors, signal acquisition tests were performed as they do not present any major risks. Alongside connecting and disconnecting the electrodes, acquiring signals would help to validate the stability during the movements.

As the acquisition system was still a prototype, some adjustments had to be done for the cables (not the connector) to stay in contact with the conducting tracks and not compromise the quality of data transference. Some adjustments were made in order for a P2 male connector to be integrated with the cables.With this, the stability of the electrodes with the board was ensured without compromising signal quality, although this method may input some noise into the signal.

For this test, the electrodes were placed above the *vastus medialis* muscle, localized near the knee [63]. The subject was seated with the leg suspended and relaxed at the beginning of the acquisition, then, the person would lift the leg till its maximum contraction and remain in this position for five seconds after which would return to the resting position and repeat the process.

Before testing the proposed electrodes, some acquisitions were made using commercial self-adhesive electrodes with snap button connectors to provide data for comparison. The procedures were the same for both the commercial and the proposed electrodes except for the adaptations needed by the different connectors.

The first snap button developed had the problem of being plain and, besides falling, presented a bad connection with the female snap button connector. While the second version of the snap button was not ready for tests, the first one was used together with aluminum foil around the pin to increase the thickness and improve its contact of it with the female connector. After the second version was ready, the use of aluminum foil became unnecessary since the pin was designed following the shape of the commercial electrode but adapted for the internal dimensions of the female connector.

As the cable used to transfer the signal between the electrodes and the acquisition system had the female snap button connector for standard, the electrode with brackets had to have an auxiliary wire connecting it with the cable. Another option was to attach the wire fixed to the electrode directly to the acquisition board but it was not as stable as the first option. Figure 3.10 allows the comparison between the resulting acquired signal for each situation described.

The golden standard Ag/AgCl wet electrode results in a low noise signal with clear distinctions between the resting and contracting periods. Due to the irregular interface of aluminum foil the amount of noise in the signal increased but the stability was guaranteed, the lack of signal saturation is an indicative of it. On the other hand, the second model was developed with a higher pin what caused the occurrence of contact loss between the female connector and the upper surface of the electrode, generating the saturation showed. The connector with bracket kept the stability but the quality of the acquired signal was highly compromised since the deposition of thin film was not uniform. This is proven by the amount of resting noise present in the signal.

With the uncertainty of whether the connectors were safe or not, to test their stimulation capabilities, a resistor of $22K\Omega$ was attached to the stimulation system and the response was monitored with the oscilloscope. For this test, the snap button connector was tested using the female connector, but with alligator clips as well since damages to the thin film could cause

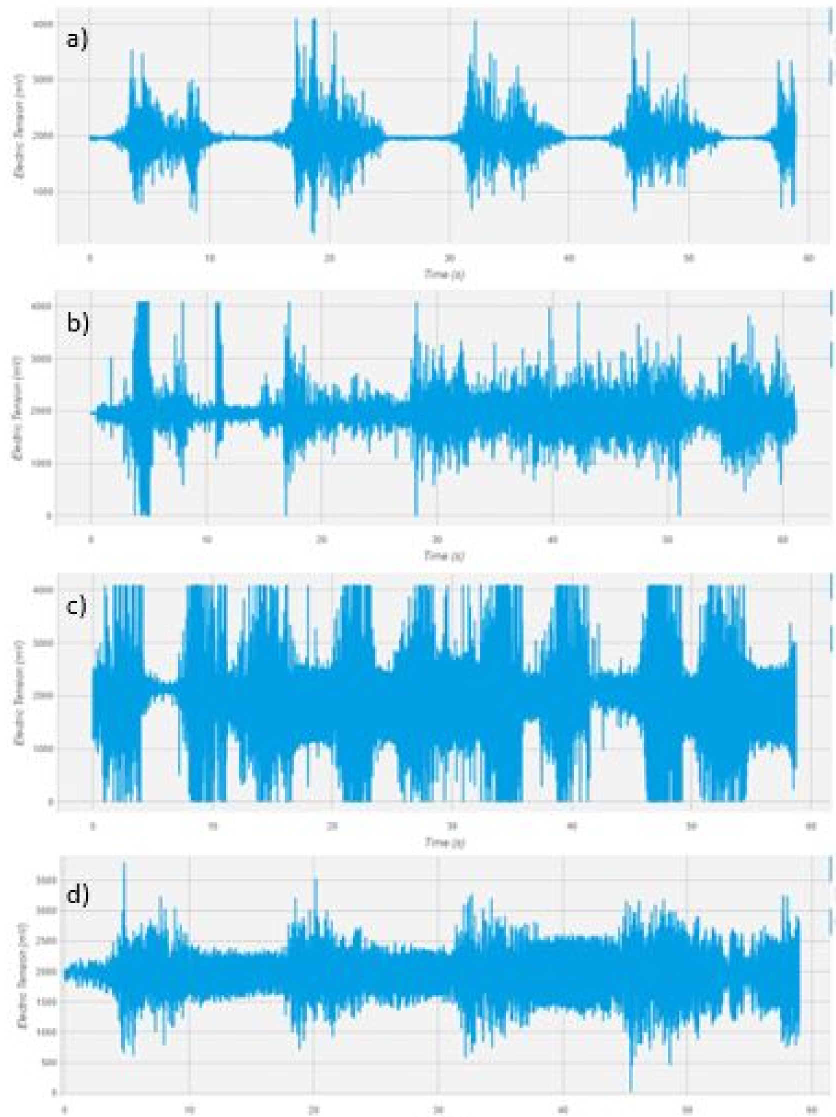

Figure 3.10: EMG results using (a) wet commercial electrodes for reference, (b) first snap button model with aluminum foil, (c) second snap button model and (d) connector with bracket

the loss of connection between the electrode and the female connector resulting in electrical shocks.

To ensure the process would not cause harm to the skin, new electrodes with intact thin films were used. An earth-leakage circuit was implemented at the output of the stimulation system. It is a security system with high impedance between the monitored circuit and the ground to redirect dangerous voltage values keeping them from damaging the circuit, in this case, the body. With this, stimulation tests were carried out directly on the skin.

In the same way as the acquisition tests, firstly the commercial electrical stimulation device was used with its carbon electrodes. Although these electrodes require the electrolyte to be applied, the output voltage was verified for comparison. After this, the same device was used but with the proposed electrodes, and the output voltage necessary for stimulation was compared between the wet carbon electrodes and the dry titanium based electrodes.

For stimulation, some minor imperfections on the thin film did not cause problems. Only heavily damaged connectors presented problems to conduct the stimulation pulses. However, all types of connectors were able to transmit the pulses if the thin film was not degraded, which was the case with some connectors with brackets.

Due to their geometry, the deposition could not cover some parts of the connector with bracket (see Figure 3.11). The biggest problem was that the part usually uncovered was the area where the wire should stay in contact with the thin film, therefore, the lack of contact made it impossible to transmit the pulses.

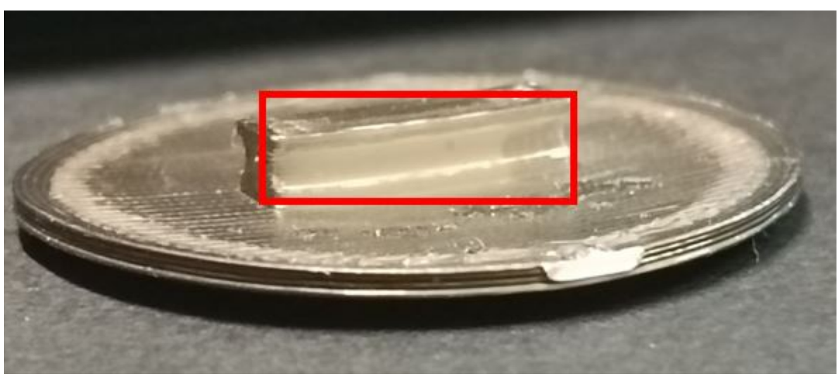

Figure 3.11: Connector with bracket with uncovered area in red

3.4 Summary

It was necessary to choose the connector type as it directly affected the quality of acquisition and stimulation. To accomplish this task, the connection and disconnection test was of great value seeing that each substrate has its features, e.g. PU had its thin film damaged due to its flexibility and cellulose broke easily. On the other hand, PLA is not as flexible as PU but not as hard as cellulose. So far, PLA has proven to be the wisest choice to be the electrode material, but more experiments needed to be done.

Since the electrode without structures had its thin film easily damaged during the tests, its implementation in the stimulation and acquisition systems would only provide reliable results for the first tries while the thin film was still intact. The connector with bracket was designed using a somewhat complex geometry making it difficult for the thin film to cover all parts equally. Besides, the bracket is not a common pattern, which would difficult the integration with other commercial components.

After the new design was implemented, the snap button connector could be connected and disconnected freely without compromising the thin film or the connector. Even if the stability continued at satisfactory levels, the increase in height made the female connector maintain contact only with the pin as there was a gap shown in Figure 3.12. Since the electrode was made through FDM, thin film deposition was not able to overcome the irregularity of the pin's surface, worsening the conductivity. Therefore, the snap button connector was the connector chosen to be implemented on the final version but the improvements remain as future work.

Figure 3.12: Last snap button version connected to the female snap connector

Chapter 4

Characterization methodology for electrodes

To safely implement the electrodes on a wearable and to choose the best option among them firstly it is necessary to know the characteristics of each different substrate and thin film. As [64] vastly describes the chemical properties of the electrodes this work will focus on mechanical and electric properties.

The possible combinations of substrate, diameter, and thin film amounts to 25 possibilities. To run all the tests on each one would require an unfeasible amount of time so the first tests to be carried out were the ones responsible to verify crucial aspects of the electrodes. With these results, some of them would be discarded in the early phases and the remaining amount would be smaller simplifying the process.

4.1 Structural tests

Before testing signal acquisition and stimulation, the electrodes should be able to withstand simpler hurdles such as the cleaning process, body fluids, and skin abrasion. I.e., if these tests were enough to incapacitate the electrodes, it would not matter if their acquisition quality was better than the others, they would not be eligible for the final product.

4.1.1 Material endurance

One of the biggest concerns about the electrodes' endurance was their integrity after the cleaning process. If cleaned by hand the possibility of damaging the electrodes and connectors is already high depending on the strength used, but inside a washing machine there are many more variables. Other clothes, shoes, and other objects may easily damage the wearable and the electrodes but as this is something hard to simulate just the outcomes of the spin cycle were taken into account.

It can vary from machine to machine so this work will consider one with a chamber of 50 centimeters in diameter and a maximum angular velocity of 1200 rpm. Converting the velocity we obtain $w = 125,7rad/s$ allowing to obtain the angular acceleration trough Equation 4.1 where r is the intern radius of the washing machine chamber, 25 centimeters in this case.

$$a = w^2 * r \tag{4.1}$$

With the angular acceleration, to obtain the maximum centrifugal force applied to the contents of the washing machine only the mass of the desired object remains. Despite being inside the wearable, the weight of the cloth does not exercise much influence on the electrodes as it is distributed in a large area. It only starts to cause significant pressure when the interleaved layers accumulate but even so it is not enough to alter significantly the force, therefore only the mass of the electrodes will be taken into account.

The average mass of electrodes with a diameter of 34mm and 15mm can be seen in Table 4.1.

Table 4.1: Electrical resistance before and after weigh test

Diameter (mm)	Number of samples	Average Mass (g)
34	8	1.045
15	12	0.27

Finally, to obtain the resultant force, mass (m) and acceleration (a) were applied in

Equation 4.2.

$$F = m * a \tag{4.2}$$

Knowing the force applied towards the electrode it is possible to simulate the spin cycle by attaching the equivalent weight to the electrodes. Firstly a weight of 300 grams was placed on top of the electrode. The small area of the 15mm electrodes brought the idea to test the pressure exercised if the weight were to be unequally distributed.

To test the pressure and make it up for the shortcomings when thinking about the whole spin cycle and the contents of the chamber the next thing was to put the weight in smaller areas aiming to increase the applied pressure. To do this, half of the electrode's body was fixed to a support and the weight was attached to the edge of the suspended half. Apart from visual changes, the electrical resistance before and after the test was verified for comparison.

With the results from the weight test and the connector tests, only PLA electrodes were subjected to the next test, thermal deformation. There were no concrete results in the revised literature about the coefficient of thermal expansion of PLA. Substrate suppliers usually have some kind of data sheet with this information but not only there were different values but also some suppliers did not include this information at all. [65] wrote a work about it but the study's subjects contained reduced quantities of PLA in its composition (20% and 40%) so the coefficients presented by *Xometry* [66] and *SD3D* [67], $41\text{x}10^{-6}K^{-1}$, and the one presented by *Simplify3D* [68] and *Cosine* [69], $68\text{x}10^{-6}K^{-1}$ were used to calculate the variation.

The first part of this test was to maintain the electrodes in a controlled temperature environment maintained at 30ºC for about 2 hours and then change them to another room at 15ºC for another 2 hours before checking if the thin film was affected. The volumetric variation ΔV, given by Equation 4.3, was calculated for this test.

$$\Delta V = 3 * \Delta L = 3 * L_0 * \alpha * \Delta\Theta \tag{4.3}$$

By using Equation 4.3, where ΔL is linear variation, L_0 its initial length, α is linear expansion coefficient and $\Delta\Theta$ is temperature variation, the theoretical change in volume was verified. The real meaning of this test was to analyze if the thin film would endure the dilatation and contraction. This could be verified by taking notes about the electrical resistance before and after the test as well as using these electrodes for signal acquisition and stimulation and comparing the results with unused ones.

4.1.2 Skin and electrode abrasion

One of the most important things to consider while dealing with the electrodes was the reactions they could cause to the skin but the damage the skin and sweat could cause to the electrode was also something important to take into account.

To prepare the skin for the session the hair must be removed and, before the electrodes are placed, the location must be cleaned with alcohol to remove the dead cells and decrease the skin impedance [70]. However, depending on how this process is carried the skin may become sensitive and in worst cases, during hair removal, it may be hurt. If on top of it the electrode presented a rough surface or burr resulting from the FDM process, the damage could be aggravated turning the session into something painful.

Since the skin preparation is supposed to be made by the patient, the only thing possible of controlling is the level of discomfort and printing quality of the electrodes. With this in mind, a simple test of fixing the electrodes in place and walking was performed. With the electrodes in place, elastic bands were used to stop them from falling while regulated to maintain the desired pressure. Then the subject had to stand up, walk for 10 minutes and remove the electrodes for analysis. This process was repeated six times with 5 minutes intervals for the analysis. At the same time, another pair of electrodes were placed on the forearm but this pair was not removed until the end of the test (90 minutes).

Some partial results were noticed but because of environmental influence on the temperature, some of these results were not recurrent such as the appearance of sweat droplets. Even so, as a corrosive fluid, the sweat was enough to cause some changes in Cu electrodes [71] being one o them shown in Figure 4.1.

The corrosion rate of the thin films threatened the patient's security and the quality of

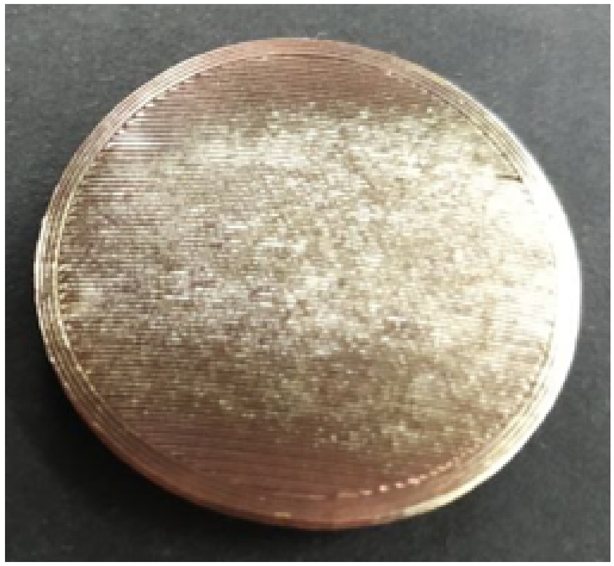

Figure 4.1: Cu electrode maintained in zip locks for five days after the comfort test

signal transmission. Therefore, to test the influence of sweat on the electrodes, physiologic sweat was prepared following the specifications provided by University of Minho (PT):

- 1 liter of de-ionized water
- 1.15 grams of Na
- 0.196 grams of K
- 0.02 grams of Ca
- 1.42 grams of Cl

One electrode of each substrate and each thin film was immersed in artificial sweat for 72 hours. The initial idea was to reproduce the experiment from [72] but after 40 hours some of the electrodes stopped showing changes. Since the other electrodes were supposed to be resistant to sweat corrosion, the experiment was previously terminated. The electrical resistance of the samples was then verified and the integrity and adherence of the thin film were tested.

4.2 Electrical resistance tests

After all the safety precautions were taken during the previous tests, and the functionality of the electrode was proven, the tests *in vivo* could take place. While validating the electrodes, stimulation system, and signal acquisition system, these tests would also provide more subjective data such as pain during stimulation and comfort.

One crucial factor tested for both stimulation and signal acquisition was the correlation between electrode placement and the quality of the response. To check it, the same parameters were used as well as pair of electrodes and other equipment while the placement was varied. The better location for the electrodes would be where the output voltage for muscle contraction was the lowest and for acquisition, where the signal reached the highest values during contraction and the lowest during rest.

The study of pressure influence on the transmission quality was carried out together with the placement test. When the best configuration of electrode placement was found, the pressure was changed by tightening or loosening the fabric mesh used for the wearable prototype shown in Figure 4.2

The circumference of the fabric mesh leggings was measured before the test and after the desired results were achieved the circumference was measured again. Clamps were used to hold the excessive cloth resulting from the pressure adjustments. With this, Figure 4.3 was consulted to verify the exercised pressure according to the perimeter.

4.2.1 Data acquisition and signal quality

To acquire EMG signals, the electrodes were placed at the *vastus medialis* and the reference was at the knee. Figure 4.4 show how the electrode was fixed with elastic bands and the rest position before the start of acquisition.

To collect data about the sensitivity of the electrodes, people with different quantities of muscular mass and fat on the leg were subjected to this test. For each one, the test was carried out with the person lifting just the leg, then the leg with a weight of 1 kilogram attached to it, and again but with 2 kilograms. Figure 4.4 also shows one of these weights attached to

Figure 4.2: Fabric mesh used for the wearable prototype being used to analyze the pressure

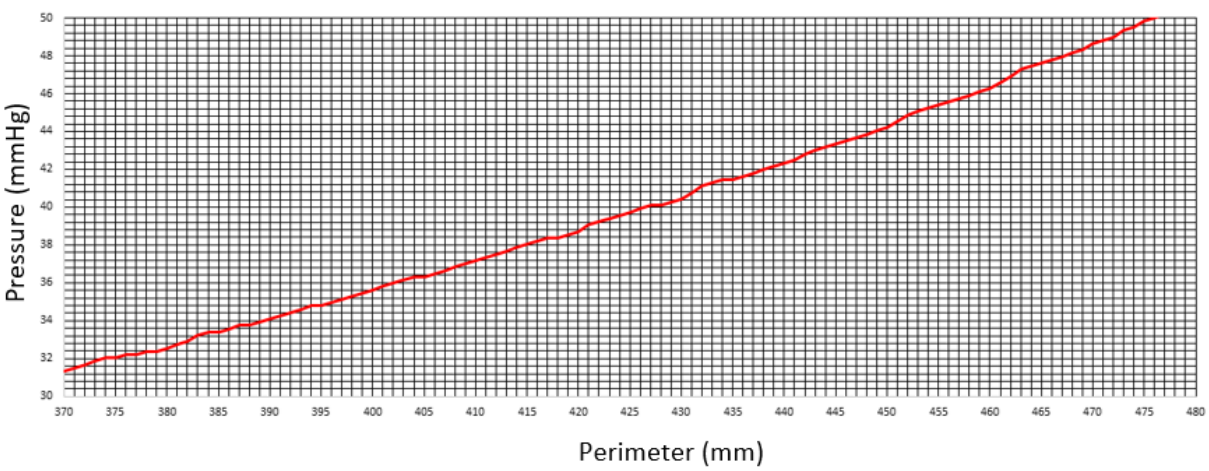

Figure 4.3: Pressure over perimeter of the fabric mesh prototype provided by *Impetus* ©

the leg. The set of movements realized was the same as the other acquisition tests, sit with the leg suspended, lift it and hold with maximum contraction for five seconds, then return to

Figure 4.4: Acquisition with electrodes fixed by elastic bands while using weight to force the muscle

rest.

4.2.2 Muscle stimulation

To test the stimulation, instead of placing the electrodes on the leg, to ease the tests and use lower voltage output, the *flexor digitorum profundus* muscle was chosen. For being localized on the forearm and being a smaller muscle, the contraction would be much more visible than on the leg as some of the fingers or the whole hand would flex according to the intensity. These involuntary hand/fingers movements would indicate that the necessary intensity was achieved or, in case of lack of movement even using high intensities, the test would be stopped and the electrodes changed for safety.

During all stimulation tests, the current was maintained at $100mA$ for safety reasons as the increase in current density towards the edge of the electrodes [73] caused by the application of square waves in disk electrodes [74] was a concern. Even with [58] affirmation that this effect is greatly reduced for electrodes with an area superior to 9 cm^2, the tests that allowed this conclusion were carried out with wet electrodes and the heterogeneous current distribution was regulated using a gel layer inside the electrode. Considering that this work aims at the implementation of thin film dry electrodes, the results achieved using wet electrodes are not applicable. Using a gel layer is also out of the scope as it would turn the proposed electrodes into semi-dry or wet electrodes.

During the study of edge effect possibilities with the proposed electrodes, it became clear that the necessary means to analyze current distribution at the surface of the electrodes would require an immense amount of time and effort. Although most studies about it were realized with wet [58] or solid metal electrodes [75] it still is an important research topic as shunting may cause harm to the skin. With this in mind, the study of this characteristic was carried out by other project participants along with deeper studies about electrical burns.

Chapter 5

Analysis of experimental results

This section summarizes the electrodes test results and clarifies their meaning. It begins by explaining the substrate choice process, followed by thin film and electrode size test outputs. To conclude the chapter, calculations about the necessary power source to supply the system are presented.

5.1 Substrate

As the objective of the electrodes is to be integrated into wearables they need to present high endurance since many people, including the medic staff and patients, sometimes handle the equipment without delicacy. Even if they do, there are much more opportunities for the electrodes to break or suffer irreversible damage. This being said, the electrodes went through an endurance test using a weight to select the best substrate for the final design.

The first part of the weight test did not result in any difference in electrode characteristics, but the second part, which consisted of fixing the electrode at the support and attaching the weight to the opposed edge did. The weight used to simulate the spin cycle during the washing process was enough to cause cracks to occur along cellulose electrodes as Figure 5.1 illustrates. The electrical resistance between the two "halves" of the electrode overloaded the multimeter as shown in Table 5.1 meaning there was no longer a connection between the thin film of the two parts.

Figure 5.1: Fissure on Cellulose electrode breaking the thin film

Table 5.1: Electrical resistance before and after weigh test

	Before		After	
Substrate	$R_b(\Omega)$	$R_{th}(\Omega)$	$R_b(\Omega)$	$R_{th}(\Omega)$
PLA	0.6	1.7	0.6	1.8
PU	0.4	1.3	278	3.1K
Cellulose	0.7	7.6	OL	OL

At the table, R_b is the electrical resistance of the electrode's flat surface and R_th is the electrical resistance between the flat surface and the pin. Table 5.1 also is of help to clarify that the flexibility of PU turns out to be a problem in some situations. During the test, PU electrodes fold instead of breaking but the thin film was not able to withstand the physical changes which resulted in an immense electrical resistance increase. The same results occurred during the connection endurance tests leaving only PLA as a viable option.

5.2 Conductivity

To characterize signal acquisition quality the tests had to be done *in vivo* as there are much more variables included when dealing with the body instead of electronic circuits. Firstly, the correct position for EMG acquisition needed to be found to not compromise the acquired data.

In this stage 7 mm electrodes were discarded as they were not even able to transmit or acquire the signal used to find the right location.

The position of the electrode has such influence on signal quality primarily because they have to be above the "start" and the "end" of the muscle or above the middle part (muscle's "belly") and the end [76][77]. Other placement configurations may cause the electrodes to acquire signals from neighboring muscles, i.e., crosstalk. The electrode's size also influences this phenomenon as the largest the electrode, the bigger the chance of it being wrongly placed and one part ending above another muscle rather than the desired one. Although [25] says that electrodes with more than 10 mm of diameter already generate crosstalk, for being made through FDM and covered with thin film, the smallest proposed electrodes capable of acquiring any useful data were of 15 mm. 34 mm electrodes were only used for stimulation as the size made them acquire too much noise.

With the locations set, acquisitions were made with PLA electrodes since PU and Cellulose had problems with endurance and connection stability, and so were excluded from the options. An example of 15 mm PLA electrode covered with Cu thin film is shown in Figure 5.2. This electrode was specifically chosen to allow better comparison with the electrode in Figure 5.3 and justify the disqualification of Cu thin film after the sweat test since it is highly susceptible to oxidation.

Figure 5.2: New and clean Cu electrode

While the Cu electrodes from each figure have a different kind of connector, the interface skin/electrode is compromised either way, which supports the objective of this test.

As Ti and $TiCu_{0.34}$ electrodes had their thin film partially removed after the same immersion test that excluded Cu electrodes, they were discarded. Acquisitions were made using these

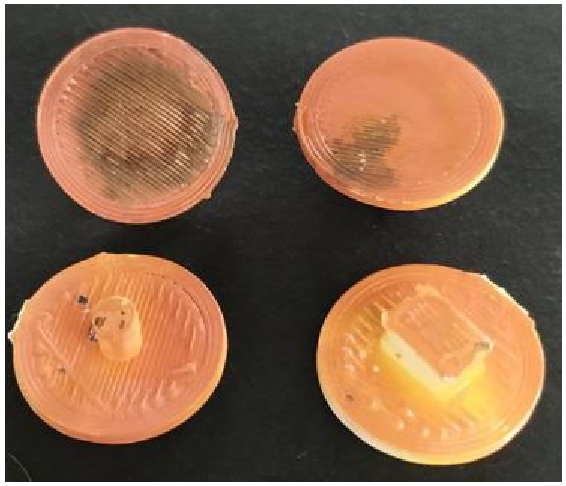

Figure 5.3: Cu electrodes highly oxidised after immersion test

electrodes as well, but as expected the absence of thin film compromised signal quality. Only TiN and $TiNCu_{0.45}$ electrodes had an unchanged appearance (see Figure 5.4) but the damages caused by sweat made themselves clear during acquisitions with TiN electrodes.

Figure 5.4: TiN electrode without apparent damages after 3 days immersed in artificial sweat

SNR was firstly used to compare the EMG signals acquired, but in some cases, it was not enough to understand if the signal achieved the necessary quality. So, Figure 5.5 illustrates the EMG signal acquired using commercial wet electrodes at the same place as the proposed dry electrodes. The only variable was the absence of elastic bands since the wet ones were self-adhesive.

As can be seen in Figure 5.6, the immersion test badly affected TiN electrodes, if not to exclude them from the options, at least to insert the need for improvements.

With the same corrosion resistance as TiN but having Cu in its composition, $TiNCu_{0.45}$ presented the best results after the immersion as can be seen in Figure 5.7. Even though, the

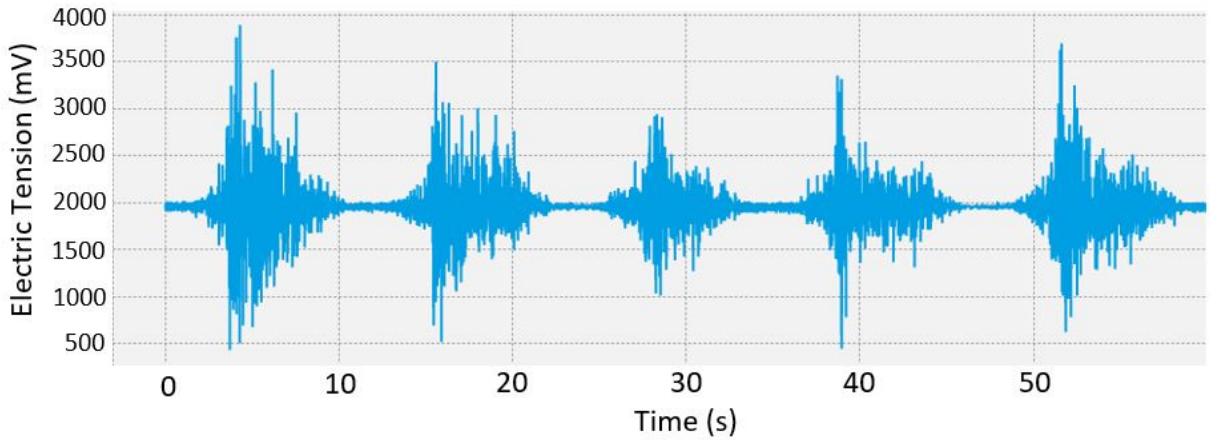

Figure 5.5: EMG signal acquired using a commercial wet electrode

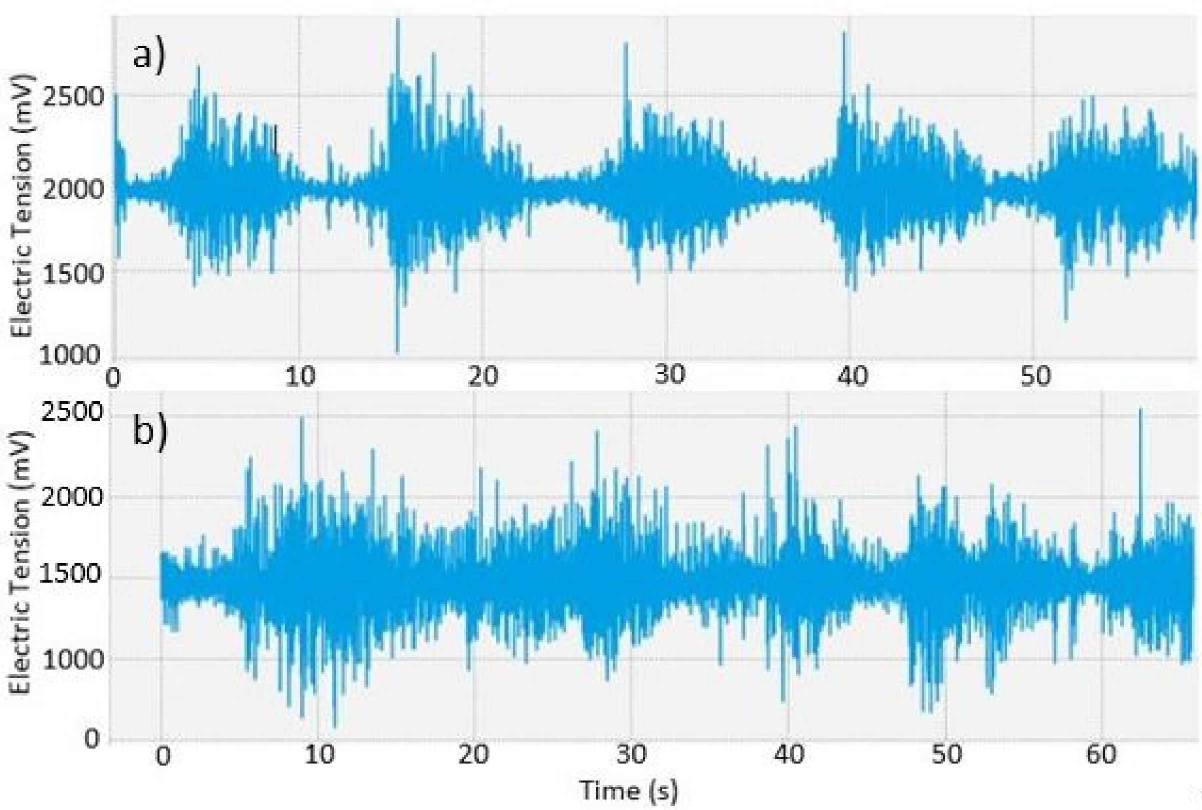

Figure 5.6: EMG signal acquired using (a) an clean TiN electrode and (b) a TiN electrode after immersion test

increase in noise still leaves room for improvements since, in some points, it hinder the muscle contraction identification.

Lastly, all the safe electrodes were used to electrically stimulate the muscle on the forearm.

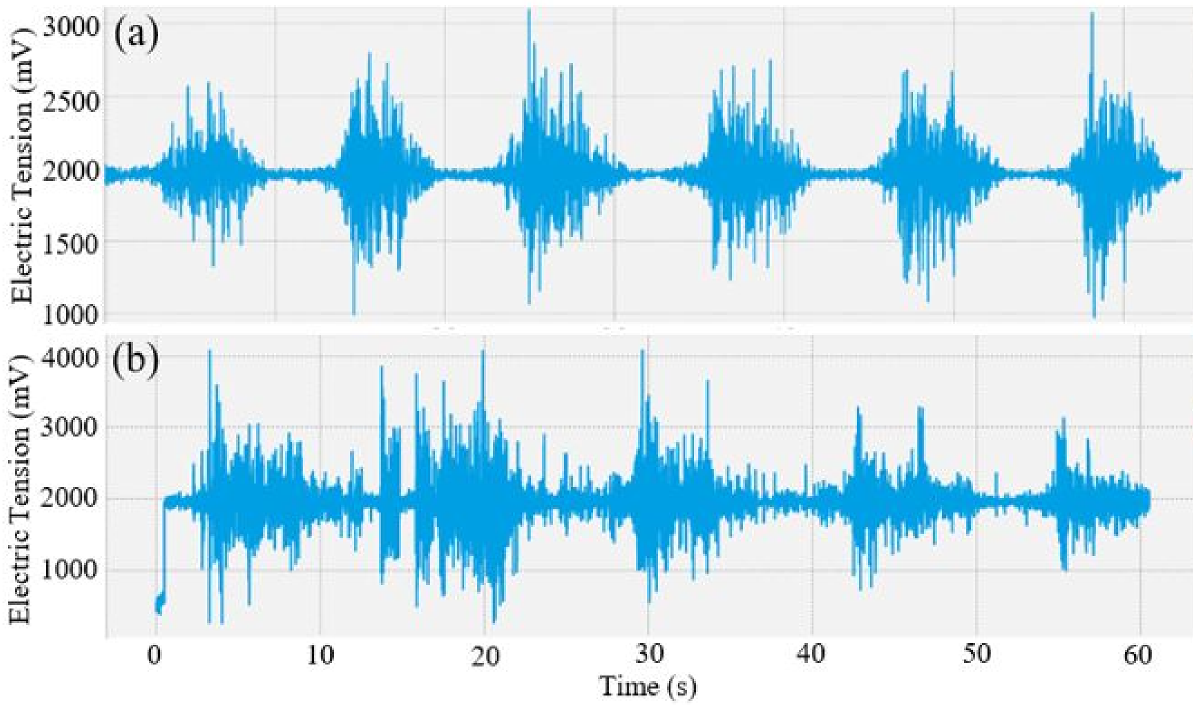

Figure 5.7: EMG signal acquired using (a) an clean $TiNCu_{0.45}$ electrode and (b) a $TiNCu_{0.45}$ electrode after immersion test

An electrode was considered to be safe if the electrical resistance never reached infinite values at any point of its surface. Even so, all the precautions were taken and with minimum discomfort, the test was interrupted. Following these requirements, the results shown in Table 5.2 were obtained using PLA electrodes since they were the best option and cellulose electrodes for comparison. PU was left out of this phase as the risk of thin film rupture was high leading to electrical shocks.

After comparing the outputs obtained and comparing them with commercial electrodes, the output voltage necessary for stimulation was higher than expected. These voltage output values were achieve using a transformer, MOSFETs and batteries. It's important to note that the skin's impedance changes according to the voltage and current levels interacting with the body. TiN electrodes along with 15 mm electrodes often caused discomfort before any sign of stimulus. The tests were stopped if the discomfort started to turn into pain or if the voltage output started to be more than 500 V.

Electrodes with diameter of 34 mm had the highest ratio of successful stimulation among

Table 5.2: Peak to peak voltage necessary for the 1^{st} contraction to be noticed durind FES

Substrate	Thin Film	Diameter (mm)	1^{st} Contraction (V_{pp})	$R_b(\Omega)$	$R_{th}(\Omega)$
PLA	Cu	34	25.3	1.9	6.3
		15	79.3	1.3	3.4
	Ti	34	124	423	13.24K
		15	106	169.3	2.6K
	$TiCu_{0.34}$	34	65	98.2	1.57K
		15	58	92.6	316.2
	TiN	34	548	13.43K	50.03K
		15	276	13.85K	2.8M
	$TiNCu_{0.45}$	34	171	5.3K	112K
		15	384	1.3K	67K
Cellulose	Cu	15	164	1.1	4.1
	$TiNCu_{0.45}$	15	412	1.8K	384K

the test subjects. Cu, $TiCu_{0.34}$ and Ti electrodes were able to stimulate the muscle without causing discomfort but were already discarded as options after the oxidation test. This left only $TiNCu_{0.45}$ as a viable option for implementation on the wearable since it withstood all tests and did not require dangerous levels of output voltage during the stimulation.

Varying system and electrode parameters had given enough results so pressure and temperature were changed to imply external factors. The variation of temperature did not increase the electrical resistance of the electrodes but in colder temperatures, the sensation of discomfort was higher and occurred for most of the electrodes with exception of $TiNCu_{0.45}$. Calculating the volumetric expansion of PLA according to Equation 4.3, the supposed options were $62.73\mu m^3$ and $104.04\mu m^3$ which were not enough to break the thin film. It is important to note that these values only apply to white PLA as other colors present different thermal dilatation coefficients [78].

For pressure changes, both the elastic bands and the fabric mesh for the wearable prototype were used. The first for easy adjusts and the second to control the pressure according to the manufacturer data. While the pressure maintained the electrodes fully in touch with the skin there were no problems with acquisition nor stimulation, but as the pressure increased both signal quality and stimulation comfort decreased. This only occurred when the pressure was high enough to start causing discomfort by itself.

In all pressure tests, there were problems keeping the electrode used as ground in place since it was placed at the knee and changed location along with the movement. To overcome this, self-adhesive wet electrodes were used to avoid floating ground of the human body [79].

5.3 Battery sizing

For the proposed wearable to be functional, the system must have a power supply integrated into it since the purpose of the wearable is to not be stuck in the same place with wires all over. To size the project's power supply, all elements that demand electric power in the system must be surveyed.

While active, the prototype had a power consumption ranging from 80 to 170 mA at 4.2V. There are three stimulation system options and, even if just one of them was used at a time, all of them were being powered, increasing the consumption. So, for the prototype to operate for 6 hours with 20% of oversizing for security reasons, the resulting power consumption will be as follows:

$$E_{prototype} = 4.2 * 0.17 * 6 * 1.2 = 5.141Wh \tag{5.1}$$

With the energy consumption of the printed circuity board (PCB), the necessary battery capacity (BC) can be obtained with 5.2:

$$BC = 5.141 * \frac{1000}{4.2} = 1224mAh \tag{5.2}$$

Although 1224 mAh already is a low value for batteries, it tends to get even lower once the stimulation system is chosen and the new PCB with just one option is implemented. Moreover, the low consumption in relation to battery capacities from commercially available batteries allows the wearable to be used for more time than expected.

Chapter 6

Conclusions and future work

The amount of existing and new physiatrists can't keep up with the growth of the elderly population, and with people having higher life expectancy this gap tends to increase more as time passes. The need to treat movement disabilities and muscular lesion is not something unusual at later stages of life so this increase in population starts to generate concerns about keeping the quality of life. Wearables that allow the specialists to monitor and set the treatments from the distance without requiring long processes emerge as a solution.

Wireless equipment with EMG signal acquisition and electrical muscle stimulation, combined with individual treatment routines represent the scope of the project. Since the idea was to develop wearables for long duration monitoring, novel dry electrodes were proposed to take the role of biosensors. Although the electrodes are responsible to stay in contact with the skin and transmit the signals, without a connector it is difficult to make them work properly. As the mediator between the microcontroller and the body, the electrodes must present a reliable connection that remains stable during the sessions seeing that a bad connection increases the noise on the acquired signal and the risk of electric shock. For this purpose, electrodes made by FDM in three different substrates, with three sizes and five thin film options were proposed. Polylactic Acid (PLA), Polyurethane (PU), and Cellulose were used as polymers to print circular electrodes with a diameter of 15 mm in PU and Cellulose cases, and with 7, 15, and 34 mm for PLA. Each combination of size and substrate was then covered by one of the five thin film options, copper (Cu), titanium (Ti), titanium-nitride (TiN), titanium dopped with copper ($TiCu_{0.34}$), and titanium-nitride dopped with copper ($TiNCu_{0.45}$).

Although the pin's width and height must be reduced to guarantee the conductibility and the stability, the current composition showed positive outcomes. The results showed that PLA electrodes, covered with $TiNCu_{0.45}$ thin film and employing snap button connector are the most suited for signal acquisition and electrical stimulation. The difference is that, for acquisition, electrodes with a diameter of 15 millimeters were chosen, while for stimulation, those with a diameter of 34 millimeters were chosen.

The technology used still has room to improve, specially the printing method. FDM is inexpensive but depending on the shape printed, the structure ends with uneven surfaces that compromise the conductivity of thin films. For this reason, the connector have a bigger role on keeping the stability than in transmitting the electrical signal as this have better results when the female connector is in contact directly with the "body" of the electrode.

Initially, the tests chosen were based on characterization methodologies found in literature reviews about dry electrodes but since the ones proposed for this work were made using a recently developed technique these methodologies weren't applicable or were time consuming. To overcome this while acquiring the necessary data, simpler tests were performed and the electrodes were used for their purpose under safe parameters for faster results.

Even if using titanium as an adhesive in the same way as electrodes found in the literature, the proposed electrodes did not present an outer layer of gold, silver, or other biocompatible and conductive material. They also are inserted in the category of film electrodes with polymer core and do not present any type of structure for contact enhancement or impedance reduction. Nevertheless, the results obtained using $TiNCu_{0.45}$ electrodes have the potential to match the signal acquisition quality and stimulation results of commercial ones without employing expensive materials or development processes.

This work should be used as basis for future works on dry electrodes characterization when dealing with substrate covered with thin films. Apart from continuing connection improvements as removing the gap present at the last version of the connector, future works include the study of other possible connections and new methods of testing its reliability. The ultimate goal is to implement and validate the electrodes with the chosen connector on the wearable and achieve commercial competitive results.

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