



Open your mind. LUT.  
Lappeenranta University of Technology

# **ELECTRICAL EQUIVALENT CIRCUITS FOR BIOSIGNAL MEASUREMENTS**

## **Biosignaalinmittausten sähköiset sijaiskytkennät**

Antti Immonen

## TIIVISTELMÄ

Lappeenrannan teknillinen yliopisto  
LUT School of Energy Systems  
Sähkötekniikka

Antti Immonen

**Biosignaalinmittausten sähköiset sijaiskytkennät**  
2018

Kandidaatintyö.

31 s.

Tarkastaja: TkT Mikko Kuisma

Hakusanat: Biosignaali, sijaiskytkentä, EMG

Biosignaalinmittauksia hyödynnetään laaja-alaisesti muun muassa lääketieteen ja puettavan elektroniikan sovelluksissa. Mitattavat signaalit ovat usein amplitudiltaan pieniä ja vaativat näin ollen huolellista elektroniikkasuunnittelua riittävän signaali-kohinasuhteen varmistamiseksi. Tässä työssä tarkasteltiin biosignaalinlähteiden sähköisiä sijaiskytkentöjä, keskittyen ihon pinnalta mitattaviin lihassähkösignaaleihin. Työn tavoitteena oli selvittää, millaisia sijaiskytkentöjä biosignaalinmittauksille voidaan muodostaa, sekä mitkä ovat kyseisten kytkentöjen tärkeimmät osat ja komponenttiarvot.

Työssä tutkittiin kirjallisuustutkimuksen menetelmin biosignaalin, analogisen etuvahvistusasteen ja ympäristön muodostamia sähköisiä sijaiskytkentöjä. Tutkimuksessa selvisi, että biosignaalin mittaaminen on haasteellista. Biosignaalinlähteille pystyy muodostamaan yksinkertaisia suuntaa antavia sijaiskytkentöjä, mutta tarkemmin mittaustilannetta kuvaavat kytkennät vaativat tapauskohtaista arviointia. Jokainen signaaliketjun osa vaatii huolellista tarkastelua, mutta tärkeimmäksi yksittäiseksi osaksi biopotentialimittausten sijaiskytkentöjä tunnistettiin elektrodirajapinnat.

Eri biopotentialimittausten tarkkojen komponenttiarvojen määrittäminen havaittiin haasteelliseksi iho-elektrodirajapinnan epävakauden vuoksi. Lisäksi erilaiset häiriölähteet kytkeytyvät helposti mittaukseen muun muassa signaalilähteen suuresta impedanssista johtuen. Puettavan elektroniikan tuomat lisävaatimukset, kuten käyttömukavuus ja pieni fyysinen koko asettavat biosignaalinmittalaitteille uusia vaatimuksia. Kuivat tai kapasitiiviset elektrodikontaktit tuovat mittauksiin sähköisiä ja mekaanisia lisähaasteita suurempien impedanssitasojen ja lisääntyneiden liikeartifaktien muodossa.

## **ABSTRACT**

Lappeenranta University of Technology  
LUT School of Energy Systems  
Electrical Engineering

Antti Immonen

**Electrical equivalent circuits for biosignal measurements**

2018

Bachelor's Thesis.

31 p.

Examiner: D.Sc. (Tech.) Mikko Kuisma

Keywords: Biosignal, equivalent circuit, EMG

Biosignal measurements are widely employed in several fields of study, including biomedical and wearable industries. The signals measured are generally of low amplitude and consequently call for careful design of the instrumentation electronics, in order to ensure adequate signal-to-noise ratio. In this Bachelor's thesis work, a literature survey was conducted on the fundamentals of the electrical equivalent circuits of biopotential signals, focusing on skin surface electromyography. The objective of the study was to introduce widely employed electrical equivalent circuits for modelling the interplay between the biosignal source, the analog front end and the surroundings. Directional component values of the most important parts were also estimated.

The results showed that directional equivalent circuits may be formed, but in order to model the behavior more accurately, case sensitive considerations are required. While all parts of the signal chain must be assessed carefully, the most important part of biopotential measurements was found to be the electrode interface. Exact component values were found to be challenging to determine, due to the dynamic nature of the skin-electrode interface. The measurements are prone to sources of noise, artifacts and interference due to the low signal levels measured and high source impedance levels often encountered. Additional requirements are introduced by wearable electronics, including unobtrusiveness and small physical size of the device. Dry or noncontact electrode interfaces introduce additional challenges in forms of increased coupling impedances and motion artifacts.

## CONTENTS

|  |    |
|--|----|
| 1. Introduction .....  | 6  |
| 1.1 Technical Framework of the Study .....                   | 6  |
| 1.2 Objective of the Study and Research Questions.....       | 7  |
| 2. Theoretical background .....                              | 8  |
| 2.1 Biophysical signal classification .....                  | 8  |
| 2.2 Biopotential signals .....                               | 8  |
| 2.2.1 Biopotential signal origin .....                       | 9  |
| 2.2.2 Surface Electromyography .....                         | 9  |
| 2.3 Electrodes .....   | 10 |
| 2.3.1 Electrode characterization .....                       | 10 |
| 2.3.2 Polarizable and non-polarizable electrodes .....       | 11 |
| 2.3.3 Capacitive properties of electrodes.....               | 12 |
| 2.3.4 Electrode placement and montages .....                 | 13 |
| 2.4 Signal amplification.....                                | 14 |
| 2.4.1 Amplifier topologies.....                              | 14 |
| 2.4.2 Input impedances .....                                 | 15 |
| 2.4.3 Generalized requirements .....                         | 15 |
| 2.5 Noise, interference and artifacts .....                  | 16 |
| 2.5.1 Noise.....   | 17 |
| 2.5.2 Interference.....                                      | 17 |
| 2.5.3 Common-mode interference and capacitive coupling ..... | 18 |
| 2.5.4 Artifacts .....  | 19 |
| 2.6 Time dependency.....                                     | 19 |
| 2.7 Noise and interference rejection .....                   | 19 |
| 3. Equivalent circuits .....                                 | 20 |
| 3.1 Biopotential signal source .....                         | 20 |
| 3.2 Skin-electrode interface.....                            | 21 |
| 3.3 Component values of the skin-electrode interface .....   | 22 |
| 3.4 EEC of a biopotential measurement system .....           | 24 |
| 3.5 Transducer EECs .....                                    | 24 |
| 4. Results .....   | 25 |
| 5. Conclusions .....   | 27 |
| References .....   | 28 |

# Nomenclature

## SYMBOLS

|          |             |
|----------|-------------|
| <i>C</i> | Capacitor   |
| <i>L</i> | Inductor    |
| <i>R</i> | Resister    |
| <i>T</i> | Temperature |

## ABBREVIATIONS

|      |                                |
|------|--------------------------------|
| ADC  | Analog-to-digital conversion   |
| AFE  | Analog front end               |
| AP   | Action potential               |
| CM   | Common mode                    |
| CMRR | Common mode rejection-ratio    |
| ECG  | Electrocardiography            |
| EDL  | Electrical double layer        |
| EEC  | Electronic equivalent circuit  |
| EEG  | Electroencephalography         |
| EMG  | Electromyography               |
| sEMG | Surface electromyography       |
| EMI  | Electromagnetic interference   |
| ESD  | Electrostatic discharge        |
| IED  | Inter-electrode distance       |
| IA   | Instrumentation amplifier      |
| MMG  | Magnetomyography               |
| MEMS | Microelectromechanical systems |
| MUAP | Motor unit action potential    |
| PSRR | Power supply rejection-ratio   |
| SNR  | Signal-to-noise ratio          |
| SoC  | System-on-a-chip               |
| TIA  | Transimpedance amplifier       |

## 1. INTRODUCTION

Recently, significant attention has been directed towards human electrophysiological monitoring, and a multitude of different sensors and gadgets are utilized [1, 2]. Smart wearables such as smartwatches and patches, are finding applications ranging from personal monitoring and entertainment to medical care [2, 5]. In light of these trends, a need arises for an electrical engineer to meet various newly realized criteria for developing this kind of physiological monitoring systems, including miniaturization, user comfort and unobtrusiveness, biocompatibility and low power consumption, to name a few. In order to be equipped to meet these requirements, one must not only understand how to design single parts of a system, but to also understand the basic operation of the entire measurement system under development, and have a grasp of the interplay between different parts of these systems, and surroundings.

This bachelor's thesis study focuses on clarifying the basics of biosignal systems from electrical engineering point of view, by means of a literature survey. Due to the long history of human biosignal measurements, there is a vast number of documentation and research papers available, but few of them create a heuristic overview of the field of study. For this reason, the thesis also includes a brief overview of the field. Based on literature sources, simplified electrical descriptions and equivalent circuits are formed to describe the relatively complex phenomena governing the formation of various biosignals in a more intuitive way. To narrow the scope of the study, this research is emphasized on the noninvasive measurements of skin-surface biopotential signals, in particular surface electromyography, sEMG.

### 1.1 Technical Framework of the Study

The field of study that deals with human biosignal measurements is diverse. A myriad of different sources of information on the human body can be harnessed to indicate the overall health and medical conditions of an individual, or to collect information about an individual's activities [1, 10]. A rough division of these sources of information can be made to depict the nature of the measured signals: electrical signals originating from the human body, and other measurable physical attributes and quantities.

This division of signal sources into biopotential and transducer signals is beneficial for understanding the signal source. In the case of endogenous biopotentials, the signal is originating from the body in forms of electric current, but in the case of transduced quantities, a conversion using a transducer has to be made in order for a physical quantity to be measurable in terms of electron flow.

In order to record these signals, a measurement system has to be realized. These systems typically comprise of a signal source on the human body, connected to an analog front end of the subsequent electronics. After obtaining an analog electric signal, the signal is rendered into digital form after possible analog conditioning steps. This digitized signal is then transmitted to a computer for further processing and analyzing, preferably wirelessly. At each of these stages, details of signal can be distorted and lost, requiring careful considerations at each stage. A simplified measurement system's front end, as well as the scope of the study are illustrated in figure 1.1.

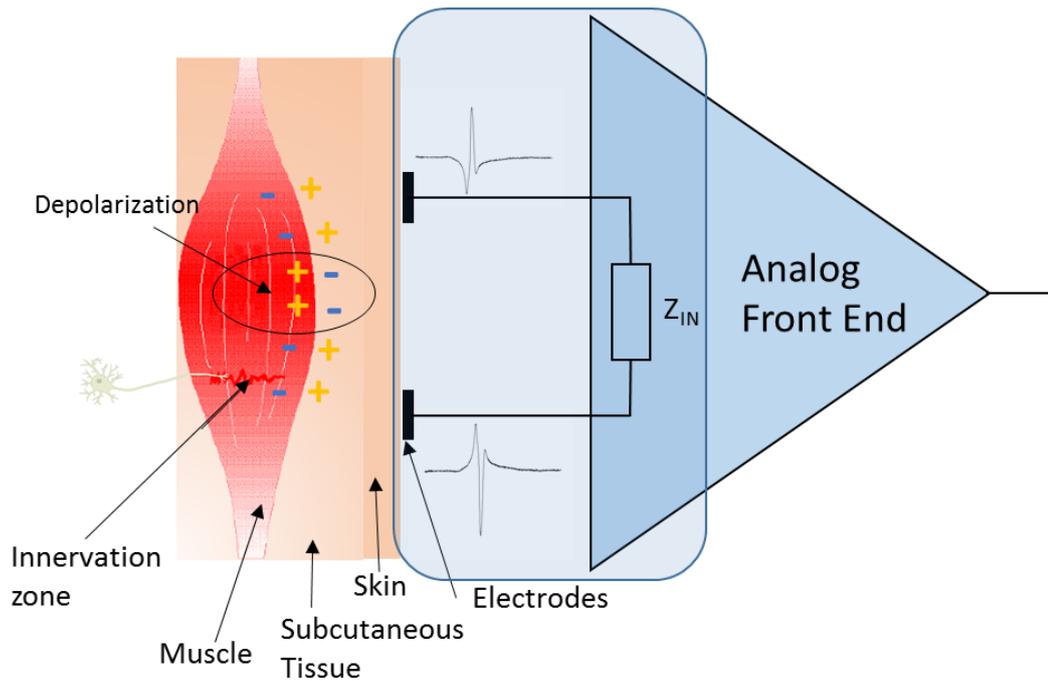


Figure 1.1 A graphical representation of a typical biopotential recording device's front end, with the scope of the study highlighted. In this study, only the equivalent circuit of the signal source and the transmission path up until the input impedance of the analog front end are considered.

## 1.2 Objective of the Study and Research Questions

The objective of this Bachelor's thesis work is to:

- Introduce important parts of equivalent circuits presented in literature, used for modelling of biophysical signal sources, with primary focus on biopotentials
- Based on sources found in literature, to develop directional, simplified equivalent circuits with approximate component values for the design of biosignal measurements system, including the signal source, the measurement instrumentation's analog front end, and rough estimates on the effect of the surroundings.

The study questions of the thesis are the following:

- What kind of equivalent circuit models are used to model biosignal measurement systems, and which are the most critical parts of these models?
- What kind of component values do these equivalent circuit elements have?

## **2. THEORETICAL BACKGROUND**

For one to be able to design a biosignal measurement system, a basic understanding needs to be obtained about the entirety of the measurement process. The separate parts of a measurement system should also be considered as an entirety, where various parts of the system might have complex interplay with one another, thus affecting the overall quality of the signal [4]. To canvas the theoretical background of the study, basic background information related to recording of biosignals is presented in this section.

### **2.1 Biophysical signal classification**

Biopotential signals originate from the body in forms of ionic current, whereas transducer signals are other forms of physical phenomena that are converted into electric signals using transducers [1]. In the case of both the transducer and biopotential types of biosignals, instrumentation usually requires high precision and gain, with high signal-to-noise ratio. It is thus important to consider the source of the signal carefully, as well as the coupling mechanisms for recording.

Transducer signals represent in number the greater variety of signals: pressure, force, strain, temperature, flow, volume, position, velocity, acceleration, rotation angle, blood chemistry, impedance, conductance, optical transmission, inclination, gas concentration, humidity, weight, surface area and density are all quantities measurable by a biophysical transducer, to name a few [10]. It is also possible to measure a single property of the body in several different ways, for example, the pulse can be measured at least electrically, mechanically, thermally, and optically.

### **2.2 Biopotential signals**

Biopotentials are signals originating from ion transfer inside the body. Biopotentials include signals originating from the muscles, electromyography (EMG), the brain, electroencephalography (EEG), and the heart, electrocardiography (ECG), among others [6]. These signals are inherently low in amplitude and in most situations are mixed with exogenous interference or other biosignals before being measured, resulting in a signal that is actually a superposition of the electrochemical potentials of many individual biological cells. These signals are transmitted from their site of origin carried by ions, while constantly experiencing propagation losses. Hence, the further away the recording electrodes are from the signal origin, the more attenuated and distorted the signal. There are also challenges present with signal coupling to the measurement device, because there needs to be a conversion from ionic current on the body to electric current inside the instrumentation electronics.

Biopotentials can be viewed as varying electric fields inside the body, which can be translated into electronic current on instrumentation electronics. In this study, only the resistive and capacitive properties of the human body are discussed. Even though magnetic fields are also present in the human body, they have conventionally been receiving significantly less attention than the measurement of electric fields, due to their low magnitudes of femto- to picotesla [8]. The measurement of the magnetic activity of the muscles is referred to as the Magnetomyogram (MMG).

### 2.2.1 Biopotential signal origin

The biopotential signals originate from mechanical contractions of biological cells, forming as a response to a multitude of simultaneously occurring phenomena. An individual cell's electric potential is known as an action potential (AP), while the action potential of an entire motor unit is called a motor unit action potential (MUAP). An action potential is caused mainly by sodium  $\text{Na}^+$ , potassium  $\text{K}^+$ , and chloride  $\text{Cl}^-$  ions being exchanged across a cell membrane in response to ionic currents or external stimuli [17, 21]. A resting potential normally exists across the cell membrane in equilibrium, due to selective permeability of the membrane to the movements of ions. This resting potential is typically on the order of -60 to -100 mV [6, 40]. When an action potential occurs, the membrane's ion permeability temporarily changes, and a depolarization potential of up to +20 mV is reached. This potential buildup is eventually diminished, in a time window of 1 – 300 ms, returning the cell membrane back to resting potential.

These ion concentration gradients induce electric fields in their surroundings, driving a propagation of an ionic signal in the volume conductor medium [13, 14]. This signal propagation undergoes propagation losses and low-pass filtration, while being superimposed with other signals present. The remnants of the action potential are then collected at a spatially separated recording point using electrodes, resulting in an electric signal on the order of microvolts to millivolts. For a more thorough biomedical description regarding the origin and propagation of the EMG signal, the reader should refer to [3].

### 2.2.2 Surface Electromyography

Surface electromyography deals with the measurement of electric potentials that are a result of a superposition of muscle cell action potentials, caused by muscle contractions [6, 44]. Due to the volume conductor properties of the human body, this electrical signal reaches two spatially separated points on the skin at slightly different points of time, resulting in a differential voltage potential between said points. It is then possible to implement a differential amplifier to compare the potentials of these two points and determine the difference, i.e. the signal of interest [6]. Another possibility is to measure the current between two or more points on the skin using a current-sensing transimpedance amplifier (TIA) [26].

To provide practical insight about the nature of biosignals, common attributes of typical EMG signals are listed below as an example. For reference, the amplitude and frequency ranges of other commonly measured biopotentials are listed in figure 2.1.

Characteristics of unamplified surface EMG signals:

- Amplitude of the signal measured at the surface of the skin is in the range of 0.1 – 2000  $\mu\text{V}$  [1, 3, 6]
- Bandwidth 6 – 500 Hz [1, 3, 6, 11], with dominant signal energy between 20 – 150Hz [1, 3, 44].
- Amplitude of EMG signal follows a quasi-random distribution, due to superposition of MUAPs [11].

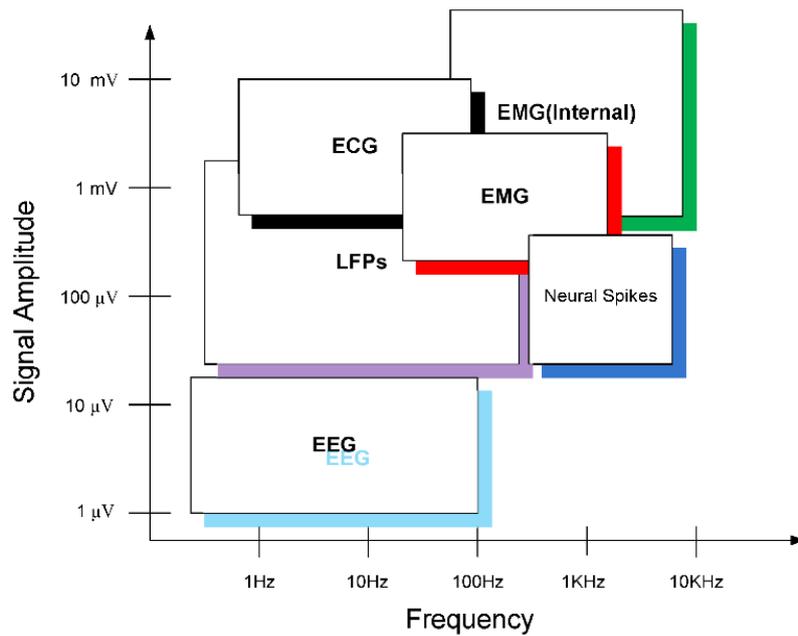


Figure 2.1 An illustration of the approximate amplitude and frequency ranges of frequently measured biopotential signals, from [7].

## 2.3 Electrodes

In a biopotential measurement system, the electrodes are a probably the most significant part of the system [15]. The importance of electrode implementation remains important despite rapidly advancing instrumentation electronics and highly integrated measurement systems such as biosignal SoCs [38]. Adequate coupling of the biopotential to the instrumentation electronics is a necessity: no matter how refined the analog front end (AFE) and subsequent electronics, if the electrodes are unreliable, then so is the entire measurement. The choice of electrode material, distance, size and shape, as well as skin preparation, amount of perspiration, relative humidity, adhesion, friction and pressure are all attributes affecting the skin-electrode contact impedance, to name a few [9, 15, 19, 27]. It is thus exceedingly important to pay close attention to the choice of electrodes and their materials, since the electrode interface is in most cases the largest contributor to measurement errors [5]. For these reasons, in this work, electrodes are paid close attention to. For a more comprehensive look at the skin-electrode interface, the effect of electrode materials and electrode noise properties, the reader should refer to [15].

### 2.3.1 Electrode characterization

Electrodes are devices that convert ionic current to electric current and vice versa. For biopotential electrodes, there exists a rough division into polarizable and non-polarizable electrodes, depending on the nature of the charge coupling mechanisms at the interface [9, 15]. In the case of a perfectly polarized electrode, the behavior is much alike that of a capacitor: little to no actual charge crosses the interface, the only current crossing being displacement current. A non-polarizable electrode, on the other hand, allows faradic current to pass relatively unhindered across the contact medium, acting only as a resistive element. Practical electrodes usually have attributes of both of these types.

Biopotential electrodes can be further divided into active and passive electrodes, depending on the location of the instrumentation electronics related to the skin-electrode interface. Passive electrodes are electrodes without active components integrated to them, and thus include long wires in between the amplifier and the electrode. This kind of electrodes often suffer from increased levels of electromagnetic interference (EMI) coupled from external sources, but are regardless widely employed due to their reliability and easily attainable low contact impedance. The most common type of clinical passive electrode is the silver/silver chloride Ag/AgCl electrode, which is a non-polarizable wet electrode. A more recent approach is to use a preamplifier attached to the electrode, thus making the electrode “active” [25].

### 2.3.2 Polarizable and non-polarizable electrodes

In the case of a non-polarizable electrode immersed in an electrolytic solution, for example a metallic electrode in contact with perspiration on skin, there are two reversible reactions present at the interface: reduction and oxidation [15]. In the case of oxidation, the anions of an electrolytic solution flow towards the metal plate, while cations are repelled from it. This is countered in the metal by a flow of negative charge carriers, electrons, away from the interface. This process is reversed in the case of reduction. This two-way process is commonly referred to as the redox reaction [6]:



where  $n$  and  $m$  are the valences of cation and anion materials  $C$  and  $A$ , respectively.

As in the case of any object exposed to a fluid, a biopotential electrode develops an electrical double layer of ions (EDL), with a thickness of up to several nanometers at its coupling medium, due to charge accumulation. [41, 15]. The double layer consists of a surface charge layer, with ions attached by chemical reactions to the surface of the material, and a diffuse layer, where the ions are loosely connected to the surface layer and undergo movement related to thermal agitation, osmotic forces, and electric charges [15]. Several models have been developed to describe the electrode interface, the first and simplest of which is the Helmholtz-model [15, 21]. A number of commonly studied electrode-electrolyte interface models are presented in figure 2.2.

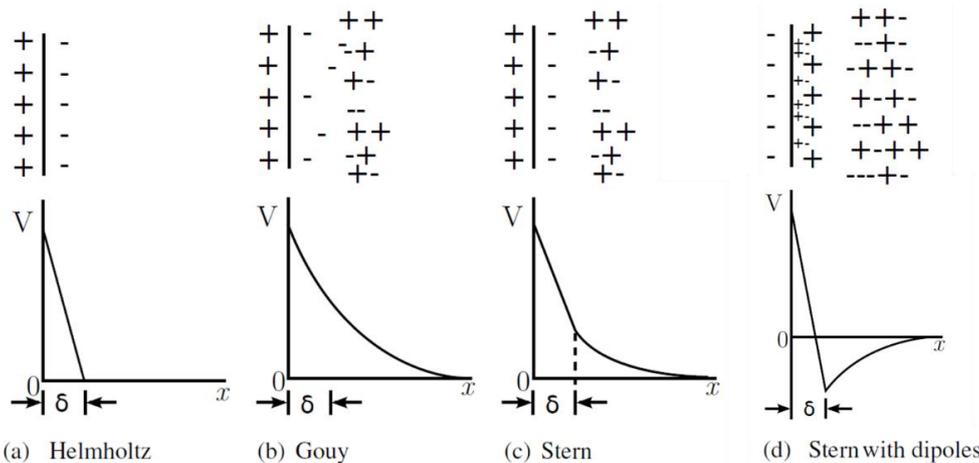


Figure 2.2 Illustrations of commonly studied electrode-electrolyte interface models in the order of their emergence, modified from [15], where  $V$  is the electrochemical potential,  $x$  is the distance spanning into the electrolyte solution measured from the surface of the electrode,  $\delta$  is the thickness of the EDL.

Another property of the electrode-electrolyte interface related to biosignal measurements is the inherent built-in electrochemical potential of the interface, the half-cell potential. This potential occurs due to ion concentration gradient near the surface of the metal, where charge neutrality cannot be maintained due to close proximity of the metal, causing the electrolyte near the metal to be at a different electrochemical potential than the rest of the solution. The half-cell potential, also known as standard potential, exhibits a distinct material dependent behavior, and the potentials compared to a standard hydrogen electrode are often compared. These standard reduction potentials,  $E_{hc}$ , are generally in the range of -1.5V to +1.5V, and can be calculated using the Nernst Equation [21]:

$$E_{hc} = \frac{-\bar{R}T}{nF} \ln \left( \frac{a_{red}}{a_{ox}} \right) \quad (2.3)$$

Where  $\bar{R}$  is the universal gas constant,  $T$  is the absolute temperature in Kelvin,  $n$  is the number of electrons transferred in the reaction,  $F$  is the Faraday constant,  $a_{red}$  and  $a_{ox}$  are the chemical activities of the reduced and oxidized parts of the ions. Tables on the magnitudes of these standard potentials can be found literature, e.g. [15].

The actual overpotential at the electrode-skin interface is usually subject to change due to various processes occurring at the interface. Passing of current through the interface, presence of electric fields, metal ion crystallization and diffusion may alter the potential from the standard potential by significant amounts, thus leading to differential potentials from one electrode in respect to another [15]. Time-dependent liquid junction potentials may also exist in between electrolytic solutions of different ion concentrations, introducing a potential on the order of tens of millivolts [16].

### 2.3.3 Capacitive properties of electrodes

Since the electrode polarization phenomenon is a property resulting from direct contact of metal and an electrolytic solution, it is possible to realize electrodes that do not exhibit quite as strongly electrolyte-dependent polarization behavior, by adding an insulator in between the electrode and an electrolyte. These kind of electrodes are known as capacitive, non-contact or insulated electrodes, where charge coupling across the interface occurs mainly by means of electric field displacement currents [5, 9].

The displacement current exchange can be heuristically envisioned by thinking of the electrode as a plate of a capacitor, the other plate being the surface of the skin. The formation of a strongly capacitive behavior even in the case of direct contact electrodes can also be explained by the fact that the surface of the skin is relatively dry. The surface layer of the skin, stratum corneum, introduces a slight barrier for ion transport, which in terms of equivalent circuit elements manifests as a series resistance in the range of kilo-ohms to megaohms in parallel with a capacitance [3, 9]. The effect of the skin surface has been conventionally reduced by skin preparation and abrasion, and use of electrolytic solutions. This kind of an approach, while providing low coupling impedance and thus excellent signal quality often introduces discomfort to the subject, especially in longer measurements. According to [9], the recent trend is the diminishing of the significance of its resistive component in favor of user comfort. Consequently, unconventional materials, such as textiles,

foams and nanomaterials are considered more frequently over the conventional Ag/AgCl type electrodes, along with capacitive electrode implementations [9, 19, 27, 29, 42]. For a thorough review of the state of art of the capacitive biopotential sensing, reader should refer to [5], and for capacitive electrode materials review, [9] can be referenced.

### 2.3.4 Electrode placement and montages

Electrode count and their spatial locations, as well as shapes and sizes also play an important role in biopotential recording. Electrode locations on skin should be chosen carefully, with both biophysical properties and user convenience in mind. Inter-electrode distance (IED), placement relative to a muscle's innervation zone, and susceptibility to movement are examples of things to consider when designing a recording setup. Recommendations on electrode placement can be found in [3] and [11].

Generally, larger inter-electrode distances result in larger signal amplitudes but with increased signal mixing and larger crosstalk. Increase in IED also increases the phase difference of the common mode signal present at the electrodes, resulting in a larger portion of the CM signal translating into a differential-mode signal [10]. Moreover, the effect of electrode size and shape can be viewed as a 2D spatial filter [3, 33]. To utilize these spatial filter properties, various electrode montages have been developed. Illustrations of some exemplary electrode montages connected to various AFE topologies are shown in figure 2.3, where IED is generally around 1 cm. For a thorough explanation about the effects of these electrode placement topologies, an interested reader should refer to chapter 5 of [3] or [43].

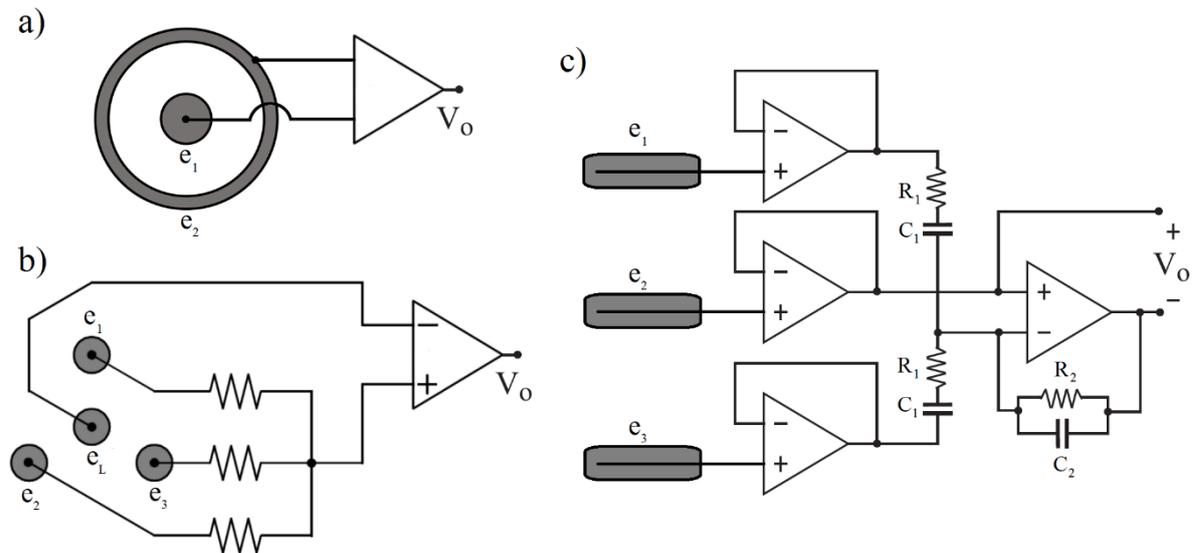


Figure 2.3 Examples of electrode montages modified from [22] and [43], contacted to their respective AFEs, with the electrode sizes, shapes and relative placements shown in grey. a) A bipolar circular electrode connected to the conventional instrumentation amplifier AFE, b) a Laplacian topology with weighing resistors [43], c) A conventional three bar electrode system with a double differential AFE [22].

## 2.4 Signal amplification

Amplifiers are necessary for increasing the signal strength of the measured biosignal with high fidelity and without disturbing the signal source [6, 17]. It is beneficial to understand the basics of the main amplifier topologies to be able to discuss the interplay between the AFE of the instrumentation circuit and the EEC formed by the human body as a signal source. However, to narrow down the scope of this study, amplifiers are only viewed as impedances loading the EECs. For better descriptions on the analog circuits related to bi-signal measurements, e.g. [17] may be referenced.

### 2.4.1 Amplifier topologies

Differential voltage amplifiers are most frequently used for biosignal measurements as general-purpose amplifiers, to amplify the potential difference between their input nodes [6]. To achieve high input impedance and CMRR, instrumentation amplifiers are generally used [12]. An exemplary practical circuit of an IA differential amplifier is shown in figure 2.4.

Instead of using a voltage amplifier, the biosignal current can be measured and converted into a voltage output, using current-to-voltage, i.e transimpedance, amplifiers (TIA) [12]. A transimpedance amplifier can be a favorable topology in biosignal measurements, due to inherent reduction of signal current attenuation by the AFE, TIA input impedance being near zero ohms. Typical applications include measurements of current from current-generating transducers, such as photodiodes, and voltage sources with large source impedance, e.g. [4]. TIAs and mixtures of voltage and current amplifiers have also been successfully implemented in myoelectric recording devices in several studies [5, 26, 37], introducing higher interference rejection and higher resilience to measurement environment challenges, such as measurements underwater. Multichannel measurements as well as implementation of complex electrode montages are slightly different to voltage amplifiers due to the effects on the low inter-electrode impedances of the system presented by the current amplifiers.

Uses of charge amplifiers and operational transconductance amplifiers (OTA) for biosignal measurements have also been reported [5]. A charge amplifier integrates input current into a voltage with a negative feedback capacitor, whereas a transconductance amplifier produces an output current from an input voltage [12]. Practical circuits for both charge amplifier and OTA are shown in figure 2.4.

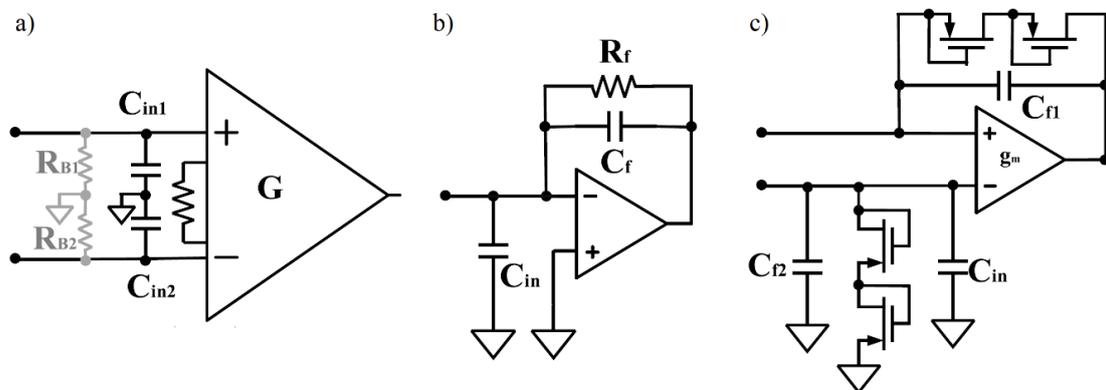


Figure 2.4 Examples of common amplifier topologies commonly used for sEMG measurements, modified from [5]. a) Conventional IA input. b) Charge amplifier, with a DC stabilization resistor  $R_f$ . A TIA can also be realized by omitting the input capacitor  $C_{IN}$ . In this case, the feedback capacitor  $C_f$  is required for AC stability. c) OTA with transistors acting as pseudo-resistors.

### 2.4.2 Input impedances

Since amplifiers are only viewed as load impedances for the signal source EECs, simplistic models for the input impedances are presented. The input impedance of an adequate differential biopotential voltage amplifier can be modelled as a resistance in parallel with a capacitance, with values of around 1 – 1000 G $\Omega$  and 1 – 10 pF, respectively [6, 3]. An OTA has likewise high input impedance, with the additional benefit of the possibility of operation in an open-loop configuration. The equivalent electrical circuit of a differential amplifier's input impedance is shown in figure 2.5.

As a directional rule, the amplifiers' input impedances should be at least two orders of magnitude larger than maximum expected source impedance, i.e. the skin contact impedance [3]. With a dry contact electrode, the skin contact impedance may be up to 1 M $\Omega$ , resulting in a requirement of 100 M $\Omega$  differential input impedance, whereas in the case of capacitive and noncontact electrodes, the required input impedance is even higher. To meet these criteria, the AFE first stages based on voltage amplifiers usually consist of a voltage follower or an instrumentation amplifier [3]. In the case of capacitive electrode systems, the desired input impedance is on the order of gigaohms to teraohms.

Another issue of the differential voltage amplifier's input is the need for a bias current path for avoiding charge accumulation induced amplifier saturation. A bias path is usually implemented by including high value resistances in between the inputs and the amplifier reference voltage, but use of such an approach has been found to introduce significant thermal noise and drift in cases where very high input impedances are desired. Various approaches have been taken to alleviate the problem, including the application of the reverse leakage current of a diode, precise design of ground plane guard rings and coupling to the next amplifier stage capacitively in unity gain topology [5]. An adequate bias path has also been accomplished by implementing a complex RC network at the input of an instrumentation amplifier, without a connection to amplifier common [35]. The need for avoiding amplifier saturation has also been compromised by periodic resetting of the amplifier. [5]

Conversely, in some cases, it can also be beneficial to make the input impedance of the AFE very low [26]. With a low input impedance, sources of high impedance current sources, e.g. photodiodes can be measured with better SNR. The input impedance of a TIA appears as an inductance of up to several nanohenries in series with a low value resistance, in contrast to the typical high impedance parallel RC circuit encountered when dealing with voltage amplifiers. The input impedance of a charge amplifier is likewise low, due to Miller effect [12].

### 2.4.3 Generalized requirements

Although most of the design parameters are dependent on the topology of the measurement's front end, a number of general requirements can be identified [1, 3, 5, 6, 17, 18]:

- **High input impedance:** minimal loading of the source is usually desired, and in the case of biopotential measurements, high input impedance has been found to help reduce the importance of electrode contact impedance magnitude and matching, and to reduce artifact and interference pickup.
- **High CMRR:** due to robust amounts of common-mode interference present in the measurement environment, the biosignal amplifier should have a high common-mode rejection ratio of at least 80 dB.

- **Low noise:** the AFE should introduce as little additional noise to the measurement as possible. Acceptable first stage noise levels are in the sub microvolt range.
- **Short distance between signal source and the AFE:** EMI pickup and signal distortion is increased with increased cable lengths.
- **Bandwidth:** The bandwidth of a biosignal amplifier is generally in the range of 0Hz to 1 kHz, depending on the measured quantity. Consequently, use of a low-pass filter at least for the purpose of an anti-aliasing filter for the ADC is recommended.
- **Patient protection:** The amplifier should have isolation and protection circuitry to ensure that there is no possibility of an electric shock for the patient. Other standards and limitations related to biopotential measurements are listed in IEC standards, e.g. IEC TC 62 and 124.

Other biosignal AFE requirements include low baseline wander, high PSRR, fast recovery from artifact-induced overload, at least moderate ESD resilience, low power consumption, small physical dimensions, ease of calibration and biocompatibility. Many of these requirements are application specific, and various tradeoffs between requirements must often be made in the development phase of the system.

## 2.5 Noise, interference and artifacts

A variety of different types of unwanted signals can seep into biomedical measurements, obscuring the signals of interest. In this section, the most dominant sources of noise and interference for biosignal electrodes and AFEs are introduced, but further analysis is outside the scope of the study. In this work, as in e.g. [16], noise is defined as an unwanted signal with roughly Gaussian amplitude distribution, whereas interference refers to external sources of more repetitive and continuous potentials. Other, more one-off type of occurrences are considered artifacts. Directional illustrations of both signals of interest and interfering phenomena are shown in figure 2.4.

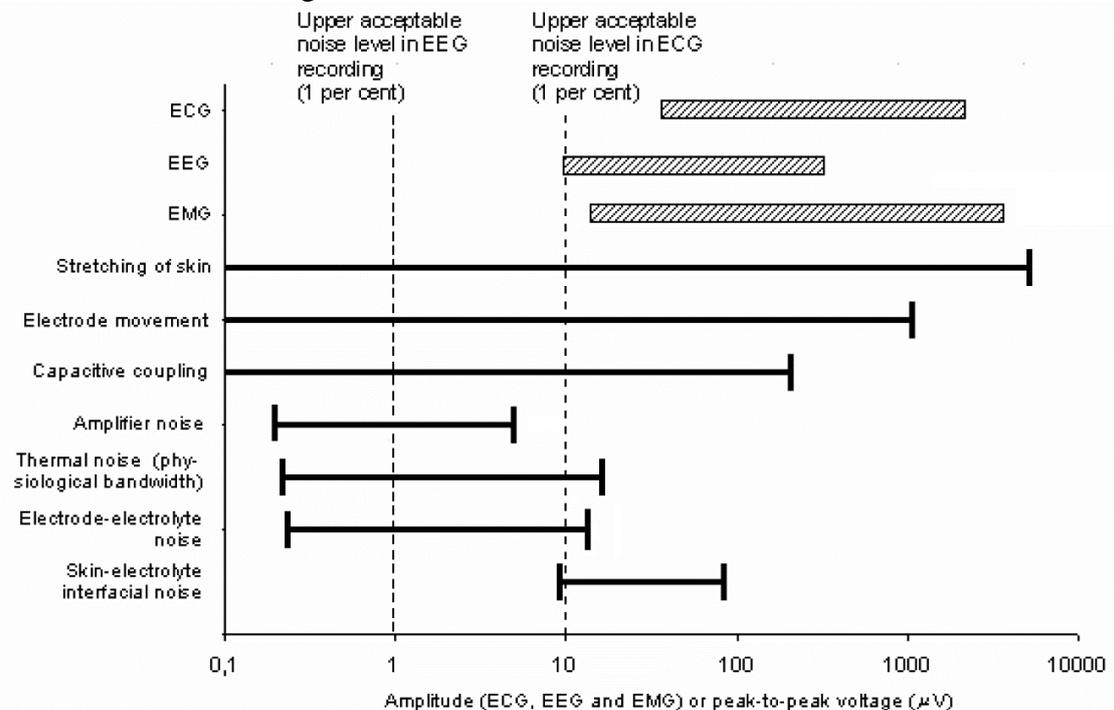


Figure 2.4 Examples of the amplitudes of typical biosignals and various sources of noise, modified from [29]. While interference and artifacts may be reduced to negligible values with careful design, sources of noise generally have nonzero minimum values.

### 2.5.1 Noise

The largest contributor to noise in a biosignal measurement is the human body, namely the skin-electrolyte interface [29]. Major sources of noise include electrode-skin contact noise, amplifier noise and thermal noise. Amplifier noise consists of several types of noise, including shot, flicker and thermal noises, manifesting in both voltage and current noises [45]. Current noise is only translated to a voltage signal when passed through a resistance, thus being more problematic when source resistance is high. Due to irregular waveforms, total noise contribution is generally quantitatively evaluated in terms of RMS summation of the individual sources.

Thermal noise is an intrinsic property of all resistive elements at temperatures above 0K, caused by thermal fluctuations of charge carriers [16, 29]. All resistive components therefore contribute to the overall noise of the measurement. Electrode-skin contact noise also consists partly of thermal noise, but is also caused by various electrochemical processes [15, 16, 29]. Electrochemical noise of the electrodes is caused by variations of the half-cell potentials  $E_{HC}$  due to presence of impurities at the electrolyte interfaces. These impurities are often conductive, and cause sudden short circuits of minuscule metal-electrode-metal cells, which in turn generate noise currents. The choice of electrode and electrolyte greatly affects the electrochemical noise at the interface: clean, gelled electrodes usually exhibit low amounts of electrochemical noise, whereas dry electrodes e.g. conductive polymers are usually more prone to noise. The largest contributor to noise in the skin-electrode interface is the electrolyte-skin contact, where a large portion of the noise is due to thermal noise proportional to the resistive component of the contact impedance [16]. The magnitude of electrochemical noise is generally around  $10\mu\text{V}$  RMS in the frequency range of biopotentials [29].

In addition to electrochemical and thermal noises, the instrumentation electronics also contribute to the overall noise. Amplifiers contribute to the noise of the measurement system by adding to the thermal noise contribution, and by electron-hole recombination processes, resulting in flicker and shot noises [15]. Separate current and voltage noise components can be distinguished, where the effect of current noise may be reduced by decreasing the source impedance through which it flows [45]. Some components of the noise generation may also be reduced by chopper-type amplifier topologies.

### 2.5.2 Interference

Sources of interference are mainly external e.g. nearby power lines, with the exception of other bioelectric phenomena such as EMG crosstalk. The human body readily acts as an antenna, with continuous inundation of electromagnetic radiation [39]. Interference coupling mechanisms can be divided into capacitive, inductive, conductive and radiative coupling [34], all of which are involved to some degree in biosignal measurements. In many biosignal measurement cases, the capacitively coupled electric field interference is particularly detrimental. Although a portion of the power line interference can be attributed to inductive coupling, in biosignal measurement systems the coupling mechanism is primarily capacitive, especially in the case of a system with short wiring [23].

### 2.5.3 Common-mode interference and capacitive coupling

When the magnitude of an electric field or the capacitance of a capacitor changes, displacement currents occur. An unwanted circuit usually forms for the displacement currents caused by time-dependent electric fields via stray capacitances connecting the subject to the power line potential and ground. Thus, a voltage division occurs, where the impedances of the system, the ground connection impedance and the impedance of the unwanted power line connection dictate the relative amplitude of the power line voltage present on the subject. Since the coupling impedance to ground is usually at least two orders of magnitude larger than the impedance connecting the subject to the power line, the common mode voltage present on the subject is usually a hundredth or less of the mains voltage [10]. In the case of 50 Hz frequency, this power line contamination can be up to 2 volts on the subject's body, with coupling capacitances of e.g. 2 and 200 pF to power line and ground, respectively.

Due to impedance and phase mismatches on the skin and electrode-skin interface or the AFE components, a portion of the common mode signal usually transforms into a differential mode signal due to unsymmetrical voltage division. [10, 23]. A comprehensive model for power line interference can be found in [23].

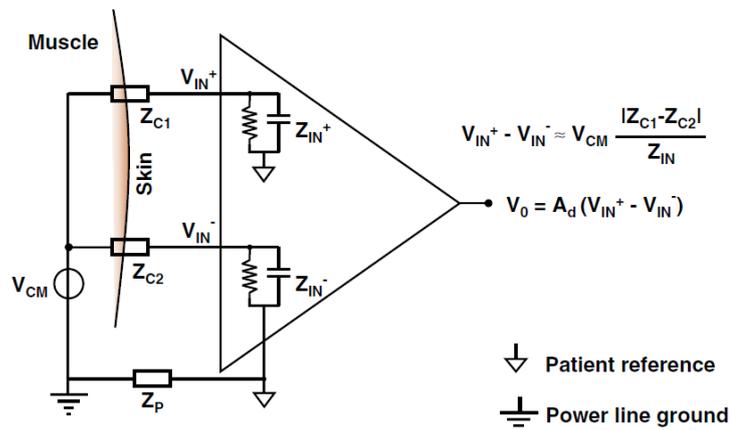


Figure 2.5 An equivalent circuit of an instrumentation amplifier connected to the patient with two electrodes, illustrating undesired common-mode coupling to power lines [20].

The emphasis of the CMRR of the AFE is usually greatly emphasized due to these robust, up to several orders of magnitude larger interference potentials usually present in the measurement. However, a portion of the signal of interest can also be viewed as a common-mode signal by the amplifier, and thus be rejected. According to [37], up to half of the signal of interest can be rejected by the AFE due to high CMRR and low inter-electrode resistance. Thus, increasing CMRR indefinitely might not help improve the SNR of the measurement, especially with large inter-electrode distances [10]. Use of mixed current and potential amplifier solutions have been proposed to mitigate this kind of phenomena, and for decreasing the crosstalk of adjacent electrodes [37]. However, maximizing of CMRR is generally viewed as a good practice.

## 2.5.4 Artifacts

Artifacts are non-recurring voltage or current fluctuations that are caused mainly by movement of the measurement setup relative to the subject. Mechanical disturbances and stretching of the skin typically give rise to disturbances of the EDL charge distribution, introducing potential swings. Other sources of artifacts include induction of triboelectric charge into the instrumentations, caused by cable movements and skin contact friction. The transients induced by motion artifacts typically follow a first order system response, where the electrode interface advances towards a state of equilibrium.

Due to their unpredictable nature and comparably high amplitudes, artifacts can be particularly detrimental to biosignal measurements especially when high time resolution data is desired. Various strategies have been taken to alleviate and compensate the effects of artifacts, including mechanical design, filtering and digital signal processing [5, 36]. Use of flexible and lightweight transducers and electrodes, in conjunction with supplementary measurements for monitoring the skin contact quality have been found to provide good framework for alleviating the motion artifacts [24, 30].

## 2.6 Time dependency

Most parameters of a biosignal measurement system are inherently time dependent. Parameter value drift can result in various measurement errors, if not compensated for accordingly. This is particularly the case with measurements with the frequency span of interest reaching DC, e.g. measurements of the thermoregulatory system, where drifting of the baseline directly affects the measurement result.

The electrode-electrolyte interface in particular exhibits strong time-dependent behavior. The coupling resistances and capacitances, DC overpotentials and chemical phenomena are all subject to drift even if no additional artifacts unsettle the system. Sweat has proved to be a large source of drift, because it affects among other things the redox-reaction, as well as thermal and electrical properties of the coupling medium. In rarer cases, component drift can result in positive outcomes, such as steadily increasing coupling capacitance in capacitive biopotential measurement systems [5].

Various measures are taken to compensate for the drift of the measurement systems and the sensor. Thermal gradient induced component value drifts can be compensated by symmetrical circuit design, and reduced by using low-drift components. Dynamic baseline recovery techniques include offset null, auto-zero, and chopper stabilization techniques, as well as compensation by signal processing methods.

## 2.7 Noise and interference rejection

Various means of alleviation of noise, interference and artifacts exist. Biosignals typically reside in a narrow frequency band of under 1 kHz, so the higher frequency noise may be filtered out using a low-pass filter. Depending on the bandwidth of the measurement of interest, and on the impedances, different sources of interference may dominate. Common-mode signals may be greatly diminished by employing a high CMRR AFE, since the interfering potentials caused by external electric fields are oftentimes mostly common mode signals. Choice of amplifier and amplifier input impedance matching with the current and voltage noises affect the noise performance [10, 45]. Comprehensive descriptions of the

coupling mechanisms and mitigation are well documented in literature, e.g.: power line interference [23], amplifier noise [45], electrode interface and noise [15, 16, 29], motion artifacts [30].

Noise and interference rejection techniques listed in literature include:

- Choice of transducer or electrode type, dimensions and material [5, 15, 16]
- Mechanical considerations [5]
- Low noise, high CMRR amplifiers [45]
- Balancing of input circuitry, AC coupling [6, 35]
- Shielding and PCB layout considerations [5, 46]
- Driven Right Leg (DRL) circuit for negative feedback for nulling common-mode voltages, usually driven by a buffer amplifier or an OTA. [15]
- Positive feedback by capacitive bootstrapping [5, 9]
- Filtering and artifact compensation circuits, which may also have feedback based on supplementary measurements, such as the galvanic skin response (GSR).
- Digital signal processing [36]

Due to highly advanced low noise front-end designs, such as biosignal SoCs [38] and dynamic instrumentation circuits [47]. Noise and interference rejection requirements have been met, for the most part, in mechanically stationary conditions. The largest issue remains the various movement related artifacts that call for innovative mechanical design and active compensation circuits with feature detection systems [5].

### **3. EQUIVALENT CIRCUITS**

In this section, simplified electrical equivalent circuits (EEC) for biopotential measurements are presented based on sources found in literature, first for the individual parts of a biosignal system, followed by connecting of the adjacent parts of the system to conceptualize a signal source, coupling path and the input impedance of an AFE. At the end of the section, a comparison to an EEC of a heat flux sensor (HFS) is included. For more comprehensive equivalent circuit models of biosignal systems, the reader is referred to e.g. [3], [31], and [40].

#### **3.1 Biopotential signal source**

As discussed in the earlier sections, the EMG is inherently an ionic signal at its origin. Ion exchange across a muscle generates electric fields that induce ionic currents in the surrounding medium. This ion current then propagates outward from the source, through a tissue of comparably low resistivity, on the order of  $500 \Omega\text{cm}$  [28]. The tissue around the source of the signal, as well as the source of the signal itself, can be viewed as a volume conductor with varying resistance as a function of frequency [3, 32]. For the sake of simplicity and to better suit the needs of simplified electrical circuit design, the ionic signal source can be represented as a dipole source inside a volume of tissue [13]. Consequently, the volume conductor properties of the subcutaneous tissue can be simplistically represented as series and parallel connected resistors. There are, in reality, among a multitude of electrochemical phenomena, capacitances and DC-potentials included in the subcutaneous layers, but those are omitted from the study. It is also important to remember that the volume conductor concept, while being of heuristic use, is also just a way to model the complex behavior of the

subcutaneous signal propagation, where both the volumes of signal source and the conductive path are anisotropic in transport properties. For a more comprehensive description on the electric volume conduction, the reader is referred to [3] and [13].

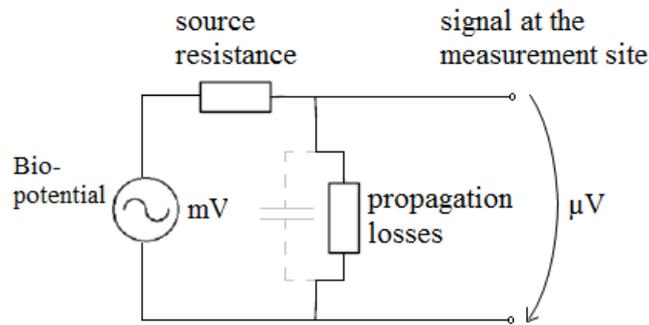


Fig. 3.1 A simplified model of the source of a biosignal along with inherent distance dependent propagation losses introduced by the volume conducting tissue.

### 3.2 Skin-electrode interface

As concluded in the background section, signal coupling at the skin-electrode interface occurs by means of faradic current through the redox-phenomena, by electric field displacement currents, or by a combination of the two. Several layers of differing ion and displacement current transport properties can be identified at the interface, and thus, the signal coupling path from inside the body to the instrumentation electronics is conventionally described as layered structure consisting of series of RC parallel components [9]. The most widely employed electrode interface models are illustrated in figure 3.2.

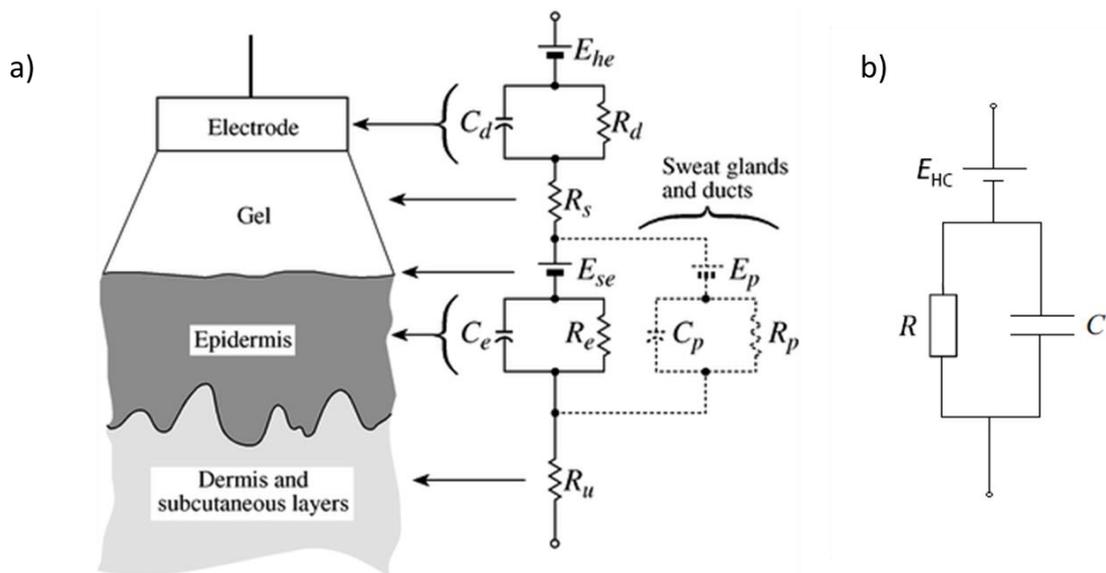


Figure 3.2 a) A widely referenced model of the skin-electrode interface found in [1]. Depending on the required simplicity of the model, fewer elements may be used to approximate the interface. In its simplest form, the EEC of the interface is that of b). Moreover, the overpotential  $E_{HC}$  is usually not relevant in the case of a noncontact electrode system.

In the case of a layered structure with several series connected parallel RC elements such as the model of figure 3.2a, the total coupling impedance  $Z_s$  can be calculated by summing together the impedances of individual RC elements [5]:

$$Z_s = \sum_{i=1}^n R_i \parallel \frac{1}{j\omega C_i} \quad (3.1)$$

Where  $n$  is the number of elements, and  $R$  and  $C$  are the resistances and capacitances of individual equivalent elements. Usually, in the case of nonclinical biopotential measurements, it is not necessary to consider all layers of the signal coupling path as individual parts, because one of the elements tends to dominate the impedance formation [9]. Consequently, the properties of the skin-electrode impedance are often approximated with a single lumped element model containing a resistance in parallel with a capacitance [15], [29]. Lumped element circuit element models encountered in literature usually resemble the models presented in figure 3.2.

The model depicted in figure 3.2 is usually sufficient for the purposes of designing simple systems for the recording of biopotentials, with a single dominating RC element representing the skin interface. More refined interface models are required for clinical applications and other forms of higher frequency skin electrical behavior, such as impedance spectroscopy and neuromuscular stimulation [3, 31, 33]. These models usually include higher order RC circuits and constant phase elements (CPEs) to account for various subsystems comprising of skin and tissue layers. In addition, a series or parallel Warburg RC element is usually included for modelling diffusion of charge carriers at the interface, to describe time dependent concentration "waves" with certain penetration depths in the electrolyte [15]. Models including RC circuit representations of the subcutaneous systems are of more use in the case of higher frequency studies due to lower values of capacitive reactance in the frequency span of biopotentials [32].

### 3.3 Component values of the skin-electrode interface

The component values of the EEC are generally difficult to determine. While being dependent on time, frequency, and coupled current densities, they also vary by several orders of magnitude depending on the application and various electrode parameters. Due to incremental advances in instrumentation electronics, the use of entirely capacitive electrodes is becoming more and more feasible. Along with new capacitively coupled systems, the sensing electrodes also evolve. Electrodes made of various materials including fabrics, foams and nanomaterials are emerging, with a wide range of fabrication methods. Electrical equivalent circuit models of these kind of electrodes are generally less straightforward than that of a conventional wet electrode. In [9], approximate component values for the coupling impedances have been presented for wet, dry, and capacitive electrodes. In figure 3.4, exemplary electrode contact interfaces are presented, along with directional component values.

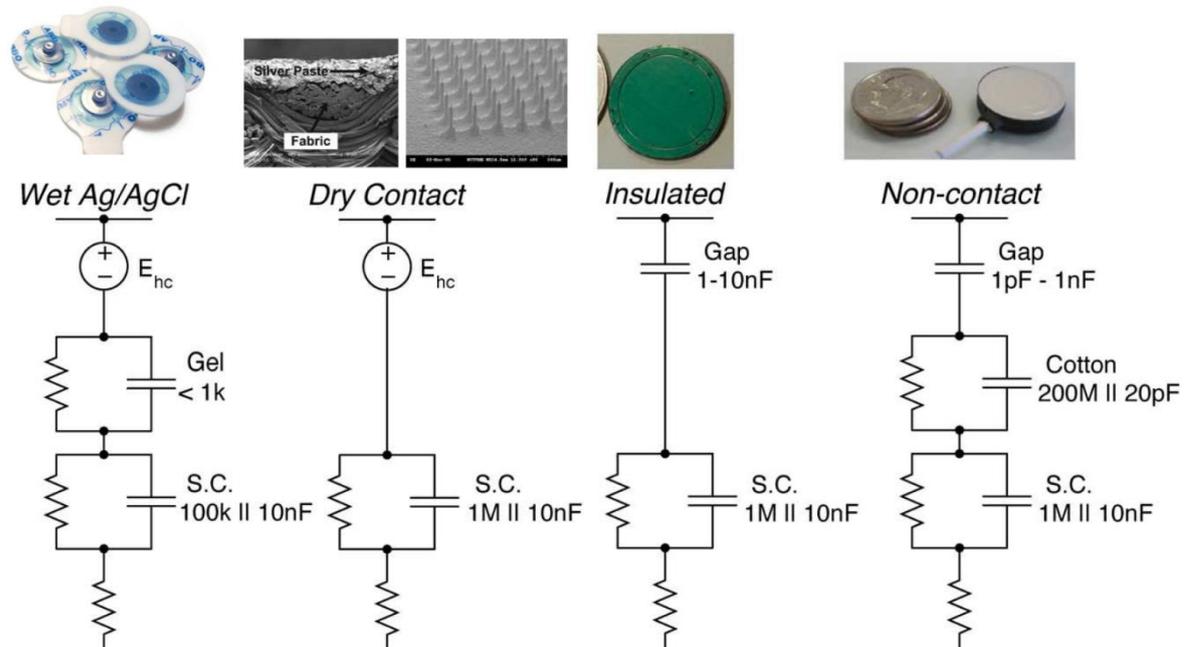


Figure 3.4 Directional component values for various electrode types. In the case of wet and dry contact type electrodes, there exists an electrolyte-related polarization overpotential  $E_{HC}$ , which does not exist with capacitive electrodes. The stratum corneum, abbreviated S.C. in the figure, typically introduces a large impedance onto the coupling path.

The impedance behavior of dry and noncontact electrodes is more unpredictable and unstable compared to that of wet electrodes, with the impedance typically being of very high value. Even in cases with electrodes isolated by fabric, in addition to capacitive coupling, there can be large series resistances present. These coupling resistances are usually on the order of 100 M $\Omega$  or larger, but can still contribute more to the contact impedance than the formed capacitance, if the size of the capacitive electrode is small or far from the skin [9]. Aside from setting the required input impedance of the amplifier high for avoiding signal attenuation, a high coupling impedance can make the system more prone to noise and interference. A high value resistive element introduces a significant amount of thermal noise to the measurement, and a high coupling impedance typically means higher amount of capacitively coupled interference. These notions generally direct the designer to favor systems with low coupling impedances. Surprisingly, it is shown in [9] that noise performance is also optimized in case of infinite coupling resistance. Due to requirements of user comfort and unobtrusiveness, capacitive biopotential systems are regardless studied in increasing amounts.

### 3.4 EEC of a biopotential measurement system

Upon connecting the aforementioned parts of a biosignal system, an EEC can be realized. The signal source can be simply modelled as a voltage source with negligibly small source resistance, whilst a pair of parallel RC elements in combination with DC voltage sources can be used to describe the skin-electrode interface in a bipolar electrode configuration. To form a practical circuit, the source and coupling impedance is connected to the input impedance of an instrumentation amplifier, which consists of balanced high-value bias-resistances and large differential input impedance of the amplifier itself. To further bring the circuit closer to a real-life application, a directional estimate of a coupling path for capacitively coupled mains interference is also included. The combined EEC is illustrated in figure 3.5.

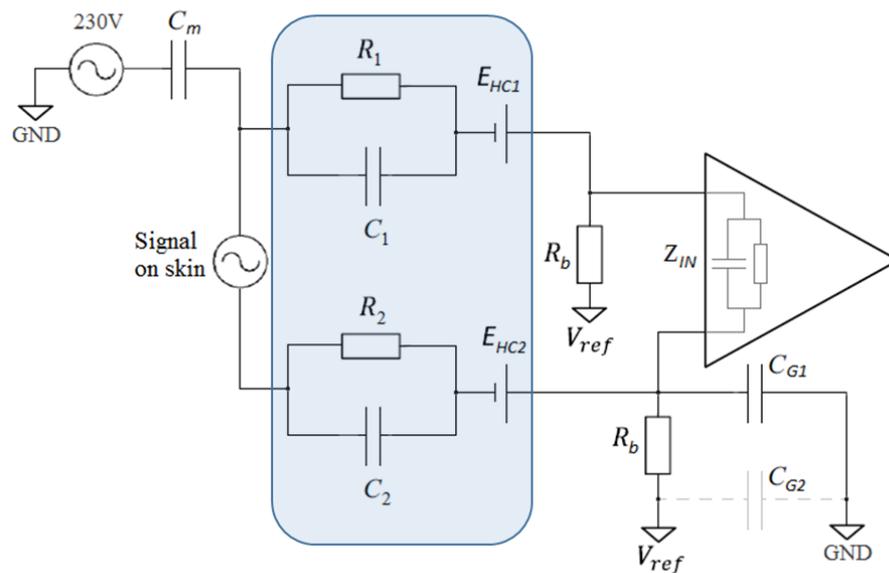


Figure 3.5 The equivalent circuit of a biosignal source and coupling impedance connected to the input of an instrumentation amplifier, with the skin-electrode interface highlighted. The common-mode interference from power lines can find several coupling paths inside the system, e.g. due to the inherently unbalanced values of the components of the skin-electrode interface.

### 3.5 Transducer EECs

In this work, the focus has been directed towards biopotential signals' EECs, but for the sake of comparison, a brief description of an exemplary transducer EEC is included. In the case of biosignal transducers, the surrogate equivalent circuit formed is in some aspects similar to that of biopotential electrodes. The transducer is located on the skin or in close vicinity of the skin, although often galvanically isolated and connected to an AFE of the measurement system. The topology of the equivalent circuit, however, depends strongly upon the type of sensor used. A division into high and low impedance transducers can be made, due to slightly different electromagnetic behaviors of different types of transducers. These differing instrumentation cases consequently require varying emphasizes on the components in the design phase. An example of a low impedance transducer is a thermoelectric device, where the internal impedance is on the order of ohms to kilo-ohms. Conversely, a photodiode is an example of higher impedance transducer, where the source impedance is high, on the order of several kilo-ohms and above.

In the typical range of interest for human thermal measurements, e.g. heat flux sensor, the source can be assumed to consist only of a voltage source with a series resistance, since the thermal phenomena on the body are usually of low frequency [6, 48], and the sensor is conventionally galvanically isolated from the subject and the surroundings. Equivalent circuit of a heat flux sensor connected to the AFE comprising of an instrumentation amplifier is shown in figure 3.6.

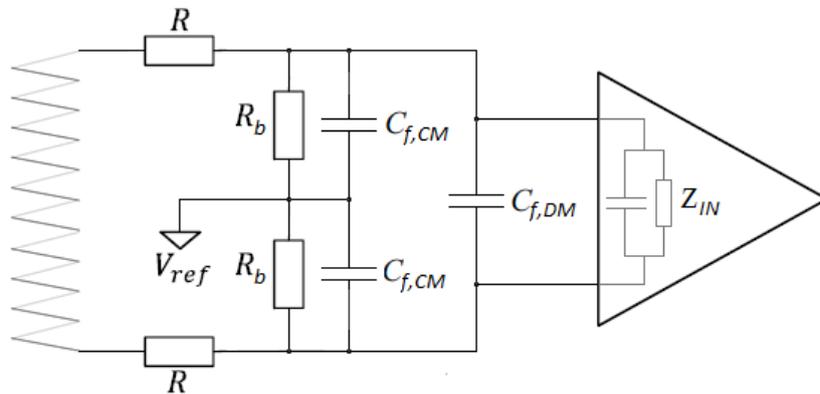


Figure 3.6 An exemplary circuit illustrating a thermopile-type heat flux sensor connected to the input impedance of the instrumentation amplifier. Due to low bandwidth of interest, the signal is low-pass filtered using resistors  $R$  and common- and differential-mode filter capacitors.

In the case of biosignal transducers, such as RTDs and thermopiles, the source resistance of the sensor is low, usually on the order of  $1\text{ k}\Omega$  or less, effectively lowering the required input impedance of the AFE. These kind of low source impedance sensors are generally less susceptible to capacitive and RF interferences, but if the physical dimensions are increased and wiring loops made, there can be interference coupled by magnetic fields. When the sensor is placed directly on skin, there can also exist resistive and capacitive coupling to skin due to close proximity of the skin. In effect, an undesired biopotential electrode may form at the sensor-skin interface, with insignificant coupling of interfering biopotential signals but with a possibility of power line interference and half-cell potential variations. It is also possible to create an unintended variable capacitor in between the skin and the sensor, with varying voltage due to changes of capacitance facilitated by changes in distance to the skin. Furthermore, various anisotropic thermoelements generally exhibit high impedance behavior, further complicating the design.

#### 4. RESULTS

Although general design guidelines and standardized practices apply, different biosignal measurement systems require slightly different emphasizes on their designs. A categorization can be made into two subgroups of signals: biopotential and transducer signals. For the purpose of nonclinical biosignal measurements, simplified lumped element equivalent electrical circuit models presented in chapter 3 were found to describe the operation of these measurement systems relatively well. However, the design complexity of a biosignal measurement system greatly depends upon the set requirements for the system, as well as the type of signal measurement to be performed. Various factors also seem to introduce tradeoffs between recorded signal fidelity and instrumentation quality, including user comfort, component cost and power consumption.

The electrode-electrolyte interface is inherently dynamic in nature, and acts as a complex impedance with capacitive and resistive components, while also including a large, time variant DC offset potential that is often at least slightly different between adjacent electrodes. For non-clinical wet electrode biopotential measurements, a sufficient equivalent circuit comprises of a signal source and a coupling impedance, connected to the input impedance of the recording device's front end. The most significant part of the biopotential signals' EEC was found to be the biopotential electrode and the electrode contact interface. When the coupling paths of typical sources of noise and interference are included, the EEC includes the elements shown in figure 3.5.

For transducer signals, the effects of surroundings can be slightly different to those of biopotential measurements. Although for the measurement of some transducer signals, such as the galvanic skin response, the EEC is very similar to that of biopotential measurements, for most transducer signals, the situation is different. In the case of low source impedance transducers such as a heat flux sensor, the effect of human proximity can be in many cases largely ignored, if adequate shielding and electrical isolation, as well as thermal and mechanical designs are ensured. The largest difference of the EEC for a biosignal transducer is the magnitude of the coupling impedance to human skin, which due to exclusion of an intentional skin contact, i.e. electrodes, is greatly increased. A generalized equivalent circuit for a thermal biosignal transducer is shown in figure 3.6.

It is often challenging estimate the component values of the EECs of biopotential and transducer signals accurately. This is due to several factors: The EEC formation and relative component values are generally highly application specific. For biopotential EECs, the type of electrode, size and shape, as well as placement topology all play a role in the formation of the equivalent circuits' component value approximations. However, several common properties have been observed in the study: In the case of surface biopotentials, the presence of a large coupling impedance introduced by the skin surface layer is usually unavoidable, although with the application of larger contact area, skin preparation and gels has been found to reduce this value drastically. The skin surface layers and the skin-electrode interface was found to play a particularly important role in the formation of the coupling impedance. The coupling impedance value is usually high, and thus the input impedance of the amplifier has to be kept high as well. Directional rules for the design of biopotential AFE design have been suggested in chapter 2. For transducer signals, many similar design recommendations apply, although the recommended AFE topology is slightly more application specific, e.g. use of a current amplifier in conjunction with high source impedance transducers. For many biosignal measurement systems, the absolute component values were not found to be necessarily very important, rather, the balance of the impedances was recognized to play a significant role.

All parts of a measurement system's signal chain have a chance of corrupting the signal of interest. While instrumentation electronics are becoming more and more advanced and integrated, the focus of design should still be kept around the entirety of the system. The importance of mechanical and materials designs seem to be likewise gradually becoming more important, to meet newly realized criteria of miniaturization, user comfort and unobtrusiveness.

## 5. CONCLUSIONS

In this Bachelor's thesis work, a literature survey was conducted on the fundamentals of biophysical measurement systems. An overview of the field of biophysical measurements was canvassed, and the typical front-end elements of the measurement system identified. Directional estimates were made about the electrical properties of the identified front-end elements, with approximate component values. Out of these elements, equivalent circuits were formed in accordance with the models found in literature for describing the basic electrical operation of typical biosignal measurements. The presented equivalent circuit models, as well as the component values are only rough estimates for giving a heuristic explanation of the basics of biophysical measurements.

The design complexity of a biosignal measurement system greatly depends upon the set requirements for the system, as well as the type of signal measurement to be performed. A recent trend appears to be the development of miniaturized unobtrusive systems in favor of bulkier, more intrusive conventional measurement topologies. To meet these requirements, several tradeoffs have to be made, including the contact quality to skin. Particularly, the source impedance plays a significant role in biotransducer instrumentation, where smaller source impedance significantly reduces EMI pickup of the system. These recent trends are reflected by new devices including e.g., MEMS and nanomaterials.

Regardless of these new trends, the basic front-end design requires attention even in the case of readily developed biosignal AFE SoCs. It is important to understand the fundamentals of underlying mechanisms of the measured quantity, so that wrong conclusions are not drawn from the measured data. Moreover, the signal properly conditioned in the AFE stage is significantly easier to post-process.

## REFERENCES

- [1] De Luca, C. (2006). Electromyography. *Encyclopedia of Medical Devices and Instrumentation*. John Wiley & Sons.
- [2] Mańkowski, T., Tomczyński, J., & Kaczmarek, P. (2015). Surface Electromyography Amplifier with High Environmental Interference Resistance. In *Progress in Automation, Robotics and Measuring Techniques* (pp. 137-145). Springer, Cham.
- [3] Merletti, R., & Parker, P. A. (Eds.). (2004). *Electromyography: physiology, engineering, and non-invasive applications* (Vol. 11). Wiley-IEEE Press.
- [4] Talvitie, J. (2015). Development of Measurement Systems in Scientific Research: Case Study. Doctoral Dissertation, LUT
- [5] Sun, Y., & Yu, X. B. (2016). Capacitive biopotential measurement for electrophysiological signal acquisition: A review. *IEEE Sensors Journal*, 16(9), pp. 2832-2853.
- [6] Webster, J. (2009). *Medical instrumentation: application and design*. John Wiley & Sons.
- [7] Saidulu, B., Manoharan, A., & Sundaram, K. (2016). Low Noise Low Power CMOS Telescopic-OTA for Bio-Medical Applications. *Computers*, 5(4), p. 25.
- [8] Garcia, M. A. C., & Baffa, O. (2015). Magnetic fields from skeletal muscles: a valuable physiological measurement?. *Frontiers in physiology*, 6, p. 228.
- [9] Chi, Y. M., Jung, T. P., & Cauwenberghs, G. (2010). Dry-contact and noncontact biopotential electrodes: Methodological review. *IEEE reviews in biomedical engineering*, 3, 106-119.
- [10] Macy, A. (2015). The Handbook of Human Physiological Recording. Referenced 7.11.2017. Available online: <http://www.alanmacy.com/HPR-Introduction.html>
- [11] De Luca, Carlo J. (2002). Surface Electromyography: Detection and Recording. Delsys Incorporated. Referenced 7.11.2017. Available online: [https://www.delsys.com/Attachments\\_pdf/WP\\_SEMGintro.pdf](https://www.delsys.com/Attachments_pdf/WP_SEMGintro.pdf)
- [12] Sedra, A. S., & Smith, K. C. (2004). Microelectronic circuits fourth international student edition. Oxford university press. *New York*.
- [13] Rodriguez-Falces, J., Navallas, J., & Malanda, A. (2012). *EMG modeling In Computational Intelligence in Electromyography Analysis-A Perspective on Current Applications and Future Challenges*. InTech.
- [14] Farina, D., & Merletti, R. (2001). A novel approach for precise simulation of the EMG signal detected by surface electrodes. *IEEE Transactions on Biomedical Engineering*, 48(6), pp. 637-646.

- [15] Riistama, J. (2010). Characterisation of wearable and implantable physiological measurement devices. Doctoral Dissertation. TUT
- [16] Hokajärvi, I. A. (2012). Electrode contact impedance and biopotential signal quality. Master of Science thesis. TUT
- [17] Northrop, R. B. (2012). *Analysis and application of analog electronic circuits to biomedical instrumentation*. CRC press.
- [18] Webster, J. G. (1998). *The measurement, instrumentation and sensors handbook*. CRC press.
- [19] Taji, B., Shirmohammadi, S., Groza, V., & Batkin, I. (2014). Impact of skin–electrode interface on electrocardiogram measurements using conductive textile electrodes. *IEEE Transactions on Instrumentation and Measurement*, 63(6), pp. 1412-1422.
- [20] Merletti, R., Botter, A., Troiano, A., Merlo, E., & Minetto, M. A. (2009). Technology and instrumentation for detection and conditioning of the surface electromyographic signal: state of the art. *Clinical Biomechanics*, 24(2), pp. 122-134.
- [21] Bronzino, J. D. (1999). *Biomedical engineering handbook* (Vol. 2). CRC press.
- [22] Guerrero, F. N., Spinelli, E. M., & Haberman, M. A. (2016). Analysis and simple circuit design of double differential EMG active electrode. *IEEE transactions on biomedical circuits and systems*, 10(3), 787-795.
- [23] Chimene, M. F., & Pallàs-Areny, R. (2000). A comprehensive model for power line interference in biopotential measurements. *IEEE Transactions on Instrumentation and Measurement*, 49(3), 535-540.
- [24] De Luca, C. J., Gilmore L. D., Kuznetsov, M., Roy, S. H. (2010). Filtering the surface EMG signal: Movement artifact and baseline noise contamination. *Journal of Biomechanics* 43. pp. 1573–1579.
- [25] Chang, K., Liu, S., Wu, X. A. (2012). Wireless sEMG Recording System and Its Application to Muscle Fatigue Detection. *Sensors*. Jan; 12(1): pp. 489–499.
- [26] Whitting, J. W., & von Tscharnier, V. (2014). Monopolar electromyographic signals recorded by a current amplifier in air and under water without insulation. *Journal of Electromyography and Kinesiology*, 24(6), 848-854.
- [27] Meziane, N., Webster, J. G., Attari, M., & Nimunkar, A. J. (2013). Dry electrodes for electrocardiography, topical review. *Physiological measurement*, 34(9), R47.
- [28] Albulbul, Ananas. (2016). Evaluating major electrode types for idle biological signal measurements for modern medical technology. *Bioengineering*, 3(3), 20.
- [29] Huigen, E. (2000). Noise in biopotential recording using surface electrodes. *University of Amsterdam Section Medical Physics*.

- [30] Cömert, A., Honkala, M., & Hyttinen, J. (2013). Effect of pressure and padding on motion artifact of textile electrodes. *Biomedical engineering online*, 12(1), 26.
- [31] Villarreal, D. L., Schroeder, D., & Krautschneider, W. H. (2010). Equivalent Circuit Model to Simulate the Neuromuscular Electrical Stimulation. *Technische Universität Hamburg-Harburg*.
- [32] Gómez-Aguilar, F., Bernal-Alvarado, J., Rosales-García, J., Guía-Calderón, M., Córdova-Fraga, T., & Sosa-Aquino, M. (2011). Frequency response of an electric equivalent circuit for a skin type system. *Revista mexicana de ingeniería biomédica*, 32(2), 93-99.
- [33] Farina, D., & Merletti, R. (2001). A novel approach for precise simulation of the EMG signal detected by surface electrodes. *IEEE Transactions on Biomedical Engineering*, 48(6), 637-646.
- [34] Ott, H. W., & Ott, H. W. (1988). *Noise reduction techniques in electronic systems* (Vol. 442). New York: Wiley.
- [35] Spinelli, E. M., Pallàs-Areny, R., & Mayosky, M. A. (2003). AC-coupled front-end for biopotential measurements. *IEEE transactions on biomedical engineering*, 50(3), 391-395.
- [36] Reaz, M. B. I., Hussain, M. S., & Mohd-Yasin, F. (2006). Techniques of EMG signal analysis: detection, processing, classification and applications. *Biological procedures online*, 8(1), 11.
- [37] von Tscherner, V., Maurer, C., Ruf, F., & Nigg, B. M. (2013). Comparison of electromyographic signals from monopolar current and potential amplifiers derived from a penniform muscle, the gastrocnemius medialis. *Journal of Electromyography and Kinesiology*, 23(5), 1044-1051.
- [38] Kim, J., & Ko, H. (2016). Reconfigurable multiparameter biosignal acquisition SoC for low power wearable platform. *Sensors*, 16(12), 2002.
- [39] Chowdhury, R. H., Reaz, M. B., Ali, M. A. B. M., Bakar, A. A., Chellappan, K., & Chang, T. G. (2013). Surface electromyography signal processing and classification techniques, review. *Sensors*, 13(9), 12431-12466.
- [40] Jobbágy, Á. & Varga, S. (2014). Biomedical instrumentation. Typotex Kiadó. ISBN 978 963 279 173 9. Referenced 12.4.2018. Available online: [http://www.tankonyvtar.hu/en/tartalom/tamop412A/2011\\_0079\\_jobbagy\\_biomedical/adatok.html](http://www.tankonyvtar.hu/en/tartalom/tamop412A/2011_0079_jobbagy_biomedical/adatok.html)
- [41] Bohinc, K., Kralj-Iglič, V., & Iglič, A. (2001). Thickness of electrical double layer. Effect of ion size. *Electrochimica Acta*, 46(19), 3033-3040.
- [42] Yao, S., & Zhu, Y. (2016). Nanomaterial-enabled dry electrodes for electrophysiological sensing: A review. *Jom*, 68(4), 1145-1155.

- [43] Fukuoka, Y., Miyazawa, K., Mori, H., Miyagi, M., Nishida, M., Horiuchi, Y., & Ueno, A. (2013). Development of a compact wireless Laplacian electrode module for electromyograms and its human interface applications. *Sensors*, *13*(2), 2368-2383.
- [44] Garcia, M. C., & Vieira, T. M. M. (2011). Surface electromyography: Why, when and how to use it. *Revista andaluza de medicina del deporte*, *4*(1), 17-28.
- [45] Texas Instruments. (2007). Noise Analysis in Operational Amplifier Circuits. Application Report. Referenced 21.4.2018. Available online: <http://www.ti.com/lit/an/slva043b/slva043b.pdf>
- [46] Microchip. (2009). Op Amp Precision Design: PCB Layout Techniques. Referenced 21.4.2018. Available online: <http://ww1.microchip.com/downloads/en/An-pNotes/01258B.pdf>
- [47] Kim, J., & Ko, H. (2016). A dynamic instrumentation amplifier for low-power and low-noise biopotential acquisition. *Sensors*, *16*(3), 354.
- [48] Childs, P. R. (2001). *Practical temperature measurement*. Butterworth-Heinemann.