Sensor-Based Garments that Enable the Use of Bioimpedance Technology: Towards Personalized Healthcare Monitoring.

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ABSTRACT

Functional garments for physiological sensing purposes have been utilized in several disciplines i.e. sports, firefighting, military and medical. In most of the cases textile electrodes (Textrodes) embedded in the garment are employed to monitor vital signs and other physiological measurements. Electrical Bioimpedance (EBI) is a non-invasive and effective technology that can be used for detection and supervision of different health conditions. In some specific applications such as body composition assessment EBIS has shown encouraging results proving good degree of effectiveness and reliability. In a similar way Impedance Cardiography (ICG) is another modality of EBI primarily concerned with the determination of Stroke Volume SV, indices of contractility, and other aspects of hemodynamics.

EBI technology in the previously mentioned modalities can benefit from a integration with a garment; however, a successful implementation of EBI technology depends on the good performance of textile electrodes. The main weakness of Textrodes is a deficient skin-electrode interface which produces a high degree of sensitivity to signal disturbances. This sensitivity can be reduced with a suitable selection of the electrode material and an intelligent and ergonomic garment design that ensures an effective skin-electrode contact area.

This research work studies the performance of textile electrodes and garments for EBI spectroscopy for Total Body Assessment and Transthoracic Electrical Bioimpedance (TEB) for cardio monitoring. Their performance is analyzed based on impedance spectra, estimation of parameters, influence of electrode polarization impedance $Z_{ep}$ and quality of the signals using as reference Ag/AgCl electrodes. The study includes the analysis of some characteristics of the textile electrodes such as conductive material, skin-electrode contact area size and fabric construction.

The results obtained in this research work present evidence that textile garments with a dry skin-electrode interface like the ones used in research produce reliable EBI measurements in both modalities: BIS for Total Body Assessment and TEB for Impedance Cardiography. Textile technology, if successfully integrated, may enable the utilization of EBI in both modalities and consequently implementing wearable applications for home and personal health monitoring.
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Borås, January 2013
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<tr>
<td>BCA</td>
<td>Body composition analysis</td>
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<tr>
<td>BLM</td>
<td>Bilayer lipid membrane</td>
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<tr>
<td>CVD</td>
<td>Cardiovascular diseases</td>
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<tr>
<td>C_{par}</td>
<td>Parasitic capacitance</td>
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<tr>
<td>EBI</td>
<td>Electrical bio-Impedance</td>
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<tr>
<td>EEG</td>
<td>Electroencephalography</td>
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<tr>
<td>ECF</td>
<td>Extracellular fluid</td>
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<tr>
<td>ECG</td>
<td>Electrocardiography</td>
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<td>EIP</td>
<td>Electrical impedance plethysmography</td>
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<td>FFM</td>
<td>Fat-free mass</td>
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<td>FRC</td>
<td>Functional residual capacity</td>
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<td>ICG</td>
<td>Impedance cardiography</td>
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<td>MF-EBI</td>
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<td>Residual volume</td>
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<td>SF-EBI</td>
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<td>SVD</td>
<td>Singular value decomposition</td>
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<td>TEB</td>
<td>Thoracic electrical bioimpedance</td>
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<tr>
<td>TLC</td>
<td>Total lung capacity</td>
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<tr>
<td>TUS</td>
<td>Tissue under study</td>
</tr>
<tr>
<td>TBW</td>
<td>Total body water</td>
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<tr>
<td>TUS</td>
<td>Tissue under study</td>
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<tr>
<td>V/I</td>
<td>Voltage / current</td>
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<tr>
<td>Z_{ep}</td>
<td>Electrode polarization impedance</td>
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<tr>
<td>SV</td>
<td>Stroke Volume</td>
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<tr>
<td>LVET</td>
<td>Left Ventricular Ejection Time</td>
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1.1 Introduction

The use of electrical bioimpedance (EBI) in the medical field has been the subject of extensive research, having attributes that make it potentially viable as a low-cost, harmless and a non-invasive technology. Diverse applications of EBI in the clinical field are being investigated and validated. Impedance plethysmography (IPG), electrical bioimpedance spectroscopy (EBIS) for total body analysis and thoracic electrical bioimpedance (TEB) for monitoring respiration rate or fluid accumulation in the lungs are some examples.

In some specific applications, such as body composition assessment, EBIS has shown a good degree of effectiveness and reliability. For example, it can effectively monitor body fluid distribution in patients undergoing peritoneal dialysis or body composition in nutrition [1-3]. Alternative methods to estimate body composition parameters such as magnetic resonance imaging (MRI) or computed tomography (CT) have shown high degrees of resolution; however, they are expensive and require highly qualified technicians.

Figure 1.1. Textile sensors integrated in functional garments utilized in the MyHeart project (Source: Philips)
In a similar way, impedance cardiography (ICG) is another modality of EBI that is primarily concerned with the determination of Stroke Volume (SV) and other hemodynamic indices. The current and most common measurement methods, including Thermodilution, Ultrasonic Doppler technique, and estimation from blood pressure, are usually costly, invasive, and require specific skills and hence, their routine use is limited to clinics and hospitals.

The use and implementation of functional garments for physiological sensing purposes is gaining market interest. The benefits of integrating electrodes into textile garments can significantly increase the development of wearable applications in different fields like Home e-Healthcare Systems. The incorporation of garment-based monitoring systems into these fields can facilitate the acquisition of physiological measurements and allow personalized health monitoring and new scenarios for point-of-care. For example, the HeartCycle initiative is an EU project intended to develop a personalized home health care system for the management and treatment of patients with Heart failure (HF), Coronary Heart Disease (CHD) and related pathologies. Daily home supervision of the patient is based on multi-parametric monitoring and analysis of vital signs and other parameters, many of which are intended to be implemented through wearable garments and some of them using EBI technology.

1.2 Motivation

Bioimpedance technology presents several benefits over traditional methods because it is considered a non-invasive, low-cost and flexible method. In addition, the incorporation of biosignal sensing garments into Home e-Healthcare Systems for multi-parametric monitoring of the patient can bring significant benefits. One of the benefits of these garments is their ability to be worn, which helps ensure and facilitate correct electrode positioning, functionality and comfort.

Measurements performed with traditional electrodes are subject to different sources of interference. These interferences are more accentuated with dry textile electrodes due to the lack of an electrolyte. Therefore, it is important to study the behavior of textile sensors so that the skin-electrode interface can be optimized and disturbances can be addressed.

1.3. Research Questions.

This research work will answer the following questions:

- Is it possible to perform Bioimpedance Measurements using textile electrodes?
- Is it possible to estimate Body Composition parameters using dry textile electrodes?
- How do the size, material and structure of the textile electrode influence impedance measurements?
Is it possible to use a sensor-embedded garment to perform an ICG analysis?

How feasible is the implementation of functional garments for multi-parametric monitoring in Home e-Healthcare Systems?

1.4. Work performed

In this research work, a study of the performance of different textile electrodes and garments for EBI measurement was done. The study was focused on the performance of textile electrodes for bioimpedance measurements in two modalities: Bioimpedance Spectroscopy (BIS) for Total Body Assessment and Impedance Cardiography (ICG) for cardio monitoring.

For the first modality, BIS for Total Body Assessment, the evaluation was based on an analysis and comparison of complex impedance spectra, Cole parameters and body composition parameters. The electrode used as reference was the Ag/AgCl RedDot electrode. Additionally important electrode characteristics such as fabric construction, skin-electrode interface, material, skin-electrode area size and design were also studied.

For the second modality, ICG, the evaluation of a set of neck and chest textrode belts for cardio monitoring was evaluated. The evaluation consisted on an analysis and comparison of ICG and ECG curves and the estimations of parameters such as SV, LVET, R-to-Z time and HR. The Ag/AgCl RedDot electrode was used as reference.

The measurements done in this research work were under the ethical approval Validering av instrument och textila sensorer för icke-invasiva fysiologiska mätningar approved by the Regional Ethical Review Board (ERB) in Gothenburg.

1.5. Structure of the Thesis Report

This work contains nine chapters and an appendix citing publications from this work. Chapter one is an introduction of the research topic and an overview of the report. The second chapter is a general review and introduction to EBI theory, measurement classification, configuration and artifacts. Chapter 3 includes potential medical applications currently using or intending to use Electrical Bioimpedance technology. It also reviews the applications studied in this research: EBIS for Total Body Analysis and Impedance Cardiography (ICG).

Chapters four and five analyze the textile theory behind a textile electrode and electrical characteristics of textile electrodes. Additional characteristics of the textiles garments used in this research are evaluated. Skin-electrode interface is studied in chapter six and Home E-Healthcare Systems, including the use of wearable technology and EBI, in chapter seven. The results obtained in this work are summarized in chapter eight. Finally in chapter nine, the obtained results are discussed, conclusions are drawn and future work is proposed.
1.6. Beyond the Scope of this Work

Motion artifacts were not studied in this work. All measurements were performed in healthy volunteers. There are several factors and conditions that may have a significant influence on the quality of measurements when using textile electrodes, including the variation of skin humidity from patient to patient and the electrode-skin contact force; however, these were not part of this study.

The conductive element in the electrode could be integrated into the electrode in different ways, most commonly with the use of metal fibers or coating with a conductive polymer. However, in this study, only the utilization of metal fibers was evaluated. Only general characteristics of the manufacturing process of the electrode such as the material, fabric construction and relevant fiber properties were considered (See Section 4.1 and 4.2). In this study, the quality characteristics of the textile electrodes, such as ease of washing and comfort, were not considered.

The evaluation of the textile belts did not aim to quantify the validity or precision of the parameters for any specific medical condition but to evaluate the performance of the belts based on the quality of the signals and the estimation of relevant parameters.
CHAPTER II

ELECTRICAL BIOIMPEDANCE

2.1. Electrical Properties of Biological Tissue

Biological tissue is composed of cells surrounded by extracellular medium. A cell is composed of several organelles or constituents enclosed by a bi-layer known as the cell membrane. The electrical properties of biological tissue are determined by the electrical characteristics of the extracellular medium, the cells and the intracellular content. The extracellular medium is an ionic solution or liquid electrolyte that is composed of diverse ions, of which, Na+ and Cl- ions are the most abundant; and the electrical properties of the extracellular medium depend on physical and chemical parameters as well as ion concentration and mobility.

Similarly, the ionic constitution and concentration of the intracellular medium determine its electrical properties. In this case, the most abundant ions are K+ and other charged molecules [4]. Although certain intracellular membrane structures have capacitive properties, the intracellular environment is generally considered to serve as an ionic conductor. The cell membrane, on the other hand, serves two roles. The first is to separate the intra- and extra-cellular media (a passive function), and the second is to control chemical exchange (active function). From the electrical point of view, the cell membrane can be considered to have very low conductive and dielectric properties. Therefore, the intracellular medium, cell membrane and extracellular medium form a conductor-dielectric-conductor structure with a capacitive behavior.

2.2. Electrical Impedance

Electrical Impedance is a measure of the opposition to electric current flow that is presented in a circuit when a voltage is applied. If the circuit carries steady direct current, current flow will be hindered by a purely resistive component created by collisions between electrically charged particles with the internal structure of the conductor. If the circuit carries alternating current an
additional reactive component will hinder charge flow. The reactance behavior is produced by changes in the magnetic and electric fields in the circuit when alternating current is flowing.

From a mathematical perspective, electrical impedance \( Z \) is derived from Ohm’s law as the ratio between voltage and alternating current. Using the complex plane in figure 2.1, impedance can be expressed in a Cartesian notation (Equation 2.1) and in a polar notation (Equation 2.2). In the Cartesian notation the impedance is represented as a complex number, where the real component represents the resistance \( R \) and the imaginary component represents the reactance \( X \). In the polar notation the impedance is represented by the magnitude of \( Z \) (\(|Z|\)), representing the ratio of voltage and current amplitude, and by an angle \( \theta \), representing the phase difference between current and voltage.

\[
Z = R + jX \tag{Equation 2.1}
\]

\[
Z = |Z|e^{j\theta} \tag{Equation 2.2}
\]

2.3. Electrical Bioimpedance Background

At the beginning of the 20th century, studies regarding the structure of biological tissue and its electrical properties contributed to the discovery that it is electrically conductive.

Further studies helped define electric circuit models that modeled the impedance and current flow of the biological tissue using capacitors and resistive bridges. The resistive component is associated with the extra- and intra-cellular fluid, while the capacitive contribution is from the cell membrane. In one of the earliest equivalent models presented by Fricke and Morse in 1925 [5], depicted in figure 2.2, the total impedance of the tissue is represented by two resistors (\( R_e \) and \( R_i \)). These represent the resistance of the extra- and intra-cellular fluid, respectively, and a
A resistor in parallel \( (R_m) \) with a capacitor \( (C_m) \) represents the cell membrane as an imperfect capacitor. According to this model, ionic current can flow around the cell in the extracellular medium and into the cell through the ionic channels across the bilayer lipid membrane BLM.

![Figure 2.2](image)

**Figure 2.2.** Fricke’s electrical model and circuit representation after a simplification. The membrane conductance usually is very small, thus \( R_m \) is disregarded.

Given this background, electrical bioimpedance can be defined as the opposition to flow of current from an externally applied electric field due to biological material. Note that the current obtained is ionic and not electronic.

As Fricke’s model suggests, the ionic current flowing through biological tissue follows different paths at different applied frequencies. As shown in figure 2.3, only a negligible amount of current passes through transmembrane channels of cells and current primarily flows in the extracellular space. However, as frequency increases, current flows through extra- and intra-cellular fluids.

Some properties of biological tissue are affected in a different manner depending on the frequency range of the current flowing through it. For instance, biological tissue has nearly constant impedance for some ranges of frequencies while it can decrease across another range of

![Figure 2.3](image)

**Figure 2.3.** Current flow in tissue at low (A) and high frequencies (B).
frequencies. This frequency-dependent behavior of bioimpedance was first identified by Schwan [6] and corresponds to an electrochemical process known as dispersion. In this process, the displacement and polarization of charges in biological tissue do not occur immediately. When the frequency is low enough, the charges have time to properly accommodate and a maximal polarization is obtained; however, when the frequency increases, polarization and permittivity are reduced.

![Diagram showing dispersion regions in biological material and approximate frequency ranges at which they occur.](image)

In figure 2.4 the dispersion regions and the permittivity in biological material are presented. According to Schwan, the major dispersions are α, β and γ, which are associated with defined frequency ranges. In these ranges, different electrochemical mechanisms take place. For clinically relevant applications, β-dispersion electrical bioimpedance is used because it reflects structural changes such as polarization of the cellular membrane and abnormal accumulation of fluid.

2.4. EBI Measurements and Classification

An EBI measurement can be classified in different ways. The two common ways of classifying the measurement in people depend on the number of frequencies utilized in the measurement and on the portion of the body that is being measured. These classifications are described in the following sections.

2.4.1. Single-Frequency Bioimpedance for BCA

Single-frequency EBI (SF-EBI) measurements for BCA are typically taken at 50 kHz with an electrode configuration known as Total Right Side, where the electrodes are placed on the right hand and the right ankle. At this frequency, the EBI measurement contains information from both the intracellular and the extracellular medium, but it is not possible to distinguish between intracellular fluid (ICF) and extracellular fluid (ECF); this approach is only possible to estimate the total body water (TBW) and the fat-free mass (FFM).
2.4.2. Multi-Frequency Spectroscopy Bioimpedance.

Multi-frequency EBI (MF-EBI) utilizes spectroscopy measurements taken between 5 kHz and 1 MHz to estimate the following BCA parameters: TBW, ICF, ECF and FFM. For this type of measurement, the most commonly used electrode configuration is Total Right Side. The multi-frequency technique has proven to be less biased and more precise than the single frequency for ECF estimation but more biased and less precise for TBW estimation [7].

2.4.3. Whole Body, Segmental and Focal Bioimpedance Measurements

Whole body bioimpedance, which provides information about the entire body, is most often performed with right-side EBI measurements and is commonly used to estimate the complete set of BCA parameters for the whole body.

EBI measurements can also be taken from different portions of the body, such as the full limb or the full trunk. These EBI measurements are called segmental EBI. If the EBI measurement is focused on a localized portion of the body, such as the belly or the ankle, the EBI measurement is called focal EBI. In focal EBI, the measurements provide information only from the specific body regions or limbs (arm, leg and trunk) of interest. Although segmental EBI is claimed to have better theoretical characterization than whole body bioimpedance, both methods produce reliable measurements [8, 9].

2.5. EBI Measurement Configuration

Because EBI is based on the passive electrical properties of tissue, energy must be applied to the tissue sample to perform a deflection measurement. Electrical energy in the form of voltage or current is applied in to the tissue under study, and the resulting current or voltage is measured. Thus, the EBI is obtained from the applied electrical energy and the tissue’s response [10, 11]. Application of electrical energy to the tissue and the measurement of the response are performed by means of electrodes.

The measurement method will determine the measurement instrumentation and thus the number of electrodes used in the system. Similarly, the number of electrodes used in the measurement system will determine the influence of the electrode polarization impedance \( Z_{ep} \) on the measurement. To illustrate the importance of \( Z_{ep} \) when performing EBI measurements, two- and four-electrode methods are introduced in the following sections.

2.5.1. Two-Electrode Measurement

In the two-electrode configuration described in figure 2.5, the same two electrodes are used for current injection and for sensing of the resulting voltage signal. Analysis of this circuit reveals how the pair of electrodes utilized to inject the excitation signal affect the measured voltage \( V_{TUS} \) (The voltage measured in the tissue under study).
Figure 2.5. Standard two-electrode configuration (Source [12])

The main drawback of this configuration is that because the electrical current flows through the sensing electrodes, the voltage generated by $Z_{ep}$ is included in the voltage measurement, and it is impossible to discern whether the voltage is generated by the current flowing through the tissue under study or by the current flowing across the skin-electrode interface. Equations (2.3) to (2.6) show the influence of the electrode polarization impedance on the calculation of the measured impedance $Z_m$ from the measured voltage, $V_m$.

$$Z_m = \frac{V_m}{I_m} \quad \text{Equation (2.3)}$$

$$V_m = V_{ep} + V_{TUS} + V_{ep} = V_{TUS} + 2V_{ep} \quad \text{Equation (2.4)}$$

$$V_{ep} = I_m Z_{ep} \quad \text{Equation (2.5)}$$

$$Z_m = \frac{V_{TUS} + 2 \cdot V_{ep}}{I_m} = Z_{TUS} + 2Z_{ep} \quad \text{Equation (2.6)}$$

### 2.5.2. Four-Electrode Measurement

In the four-electrode configuration shown in figure 2.6, the signal injection and the response measurement are performed with two different pairs of electrodes. One pair of electrodes is used to inject the current into the tissue under study (TUS), while a second pair of electrodes is used to measure the voltage $V_{TUS}$. 


With this electrode configuration, the voltage generated by $Z_{ep}$ of the injecting leads does not affect the voltage measurement. Based on equations (2.3) and (2.4), and considering that the current $I_{ep}$ flowing through the voltage-measuring electrodes is zero, the impedance measured as $Z_m$ is the impedance of the tissue under study (equation 2.7). Note that for $I_{ep}$ to be zero, the sensing amplifier must have very high input impedance, ideally infinite.

$$Z_m = \frac{V_{TUS} + 2V_{ep}}{I_m} = Z_{TUS}$$

**Equation (2.7)**

### 2.6. Measurements Artifacts

An EBI measurement can be affected by several sources of error, and as a consequence, measurement artifacts might corrupt the obtained EBI data. Although the four-electrode method is a method that removes the direct influence of electrode polarization impedance ($Z_{ep}$) on the EBI measurements, the four-electrode method does not eliminate all aspects of influence. The sensitivity to unwanted capacitances in the measurement system will increase with increasing $Z_{ep}$. These unwanted capacitances may come from stray capacitive pathways such as the capacitance between electrode leads, between body limbs and the earth, between the leads and the ground, etc. (Figure 2.7).

These stray, parasitic, capacitances form different electrical pathways so that part of the current that is supposed to flow through the TUS flows through the alternative pathways instead. The sensitivity of the measurement set-up, including the electrodes, to these unwanted capacitances is small at low frequencies.
In the circuit diagram shown in Figure 2.7, parasitic capacitances that are present in a standard EBI measurement setup are included. The capacitance $C_{le}$ represents the capacitance in the neighboring electrode leads, $C_{lg}$ represents the capacitance between the signal leads and ground, $C_{ge}$ represents the capacitance between the signal ground and the earth and $C_{bg}$ represents the capacitance between the residual body and ground.

2.6.1. The Influence of the $Z_{ep}$ on EBI measurements

Estimation of $Z_{ep}$ on EBI

As was explained in the previous section, electrode polarization impedance $Z_{ep}$ has an influence on the measurement even when the measurement has been taken using a tetra-polar configuration. Looking at the diagram in Figure 2.6, it is easy to realize that a large value of $Z_{ep}$ will most likely influence the actual value of the current injected to perform the measurement. A large value of $Z_{ep}$ will amplify any artifact related to measurement current leaking away through parasitic capacitances instead of flowing into the body through the electrode.

2.6.2. Capacitive Leakage

The Tail or Hook Effect is a measurement artifact that is noticed as a deviation in the impedance at high frequencies. The deviation is larger in reactance and phase than it is in the resistance and module of the EBI spectrum as it can be seen in [13]. The origin of this effect is the parasitic capacitances in the measurement set-up, as illustrated in figure 2.7. The existence of parasitic capacitances allows for fractions of the injected electrical current to leak away through alternative electrical parasitic pathways instead of through the measured load. Therefore, this leakage of current impairs the impedance estimation process and produces an estimation error.
Figure 2.8. Hook effect observed in A) module of the reactance vs. frequency plot and B) Cole plot.

The estimation error caused by leakage effect produces a high-frequency artifact. In figure 2.7, plots of the impedance of the TUS obtained from the model (figure 2.9) with a $C_{\text{par}}$ value of 50 pF are depicted. Impedance that is free from artifacts is plotted in blue with a solid line, and the red trace presents the impedance data that is contaminated with capacitive leakage. The reactance spectrum is plotted in A) and the impedance plot in B).

Figure 2.9. Electrical Model
CHAPTER III

ELECTRICAL BIOIMPEDANCE APPLICATIONS

3.1. Medical Applications

The dependency of EBI measurements on tissue composition and structure allows the implementation of several applications based on time analysis, single frequency and spectroscopy. EBI has proven to be an efficient and non-invasive method for patient monitoring and a diagnostic tool for several health conditions; in some applications, EBI even reached clinical implementation.

Variations in electrical impedance in the thorax are correlated with heart and respiratory activity. Impedance plethysmography (IPG) of the lungs allows the monitoring of different respiratory volumes such as residual volume (RV), functional residual capacity (FRC) and total lung capacity (TLC) for pulmonary function [14, 15]. Spectroscopy analysis of skin EBI measurements allows for early detection of melanoma [16].

Abnormal accumulation of fluid in the lungs or thoracic region, pulmonary edema, can also be detected by means of electrical impedance measurements. Commercially available devices such as the bioimpedance monitor ZOE (developed by Noninvasive Medical Technologies) measures the resistance of the thoracic region at 100 kHz to assess the accumulation of fluid in the lung [17]. Less fluid accumulation in the thoracic cavity is associated with a higher resistance, whereas low resistance values are an indicator of excessive fluid accumulation.

3.2. EBI for Total Body Composition Analysis

The use of EBI-based body composition analysis (BCA) to estimate body composition parameters has shown potential in the diagnosis and treatment of many pathologies. Water content, intra- and extra-cellular fluid and fat mass content are some of the parameters [18] that can be obtained through EBI and used in the assessment of several conditions (figure 3.1).
For instance, medical conditions such as dehydration in elderly people and in high performance athletes, over-hydration in cardiac patients with edema or lymphedema [19] and nutritional status [12, 13] can be assessed with the estimation of BCA parameters. In addition, the information provided by these parameters can be used to determine the fluid excess in patients treated with hemodialysis due to renal failure [1, 3, 20] and nutritional information in patients with Chronic Obstructive Pulmonary Disease (COPD) [21-23].

![Figure 3.1. Total body composition analysis using a four-electrode EBIS set-up](image)

### 3.2.1. Cole Function and Fitting

There are different mathematical models used to describe impedance data of skeletal muscle tissue. The Cole function is one of the models most used to describe complex impedance measurements in the β dispersion range (kHz-MHz).

In this model, the complex measurements defined by equation 3.1 form a semicircular curve with its center below the real axis. The curve presented in figure 3.2, known as a Cole plot, is a semicircular trace of resistance versus reactance, where frequency increases over the semicircle in a counter-clock wise direction. In the same plot, the resistance at zero frequency is indicated as $R_0$, whereas the resistance at infinite frequency is indicated as $R_\infty$. The characteristic frequency $F_c$, also depicted in the trace, represents the impedance at which it has a maximal dependence on the cell membrane capacitance $C_m$. 
The empirical equation 3.1, known as a Cole function, is a complex and nonlinear function of frequency presented by Kenneth S. Cole in 1940 [24]. In this equation $R_0$ and $R_\infty$ represent the resistance at the minimum and maximum frequencies, respectively, $\omega$ the natural frequency, $\tau$ a relaxation time constant and $\alpha$ a dimensionless parameter related to the spectral width of the dispersion and the morphology of the extracellular spaces [25], which takes on values between 0 and 1.

$$Z = R_\infty + \frac{R_0 - R_\infty}{1 + (j\omega \tau)^\alpha} \quad \text{Equation (3.1)}$$

Using mathematical fitting and EBIS data, Cole parameters can be estimated. An accurate estimation of the four parameters $R_0$, $R_\infty$, $\tau$ and $\alpha$ contribute to a precise estimation of body composition parameters.

Curve fitting can be implemented using different approaches and using different amounts of experimental data. Cole parameters are estimated when the Cole function is fit to the EBIS data. This fitting may be done in the impedance plane by using the properties of the semicircular Cole plot. It can also be performed on different features of the inmitance in the frequency domain. Therefore, the quality or robustness of the fitting method can be determined from several parameters such as the experimental data (Impedance $Z$, resistance $R$, conductance $G$, magnitude $Z$, etc.), number of frequencies, data distribution and fitting method.

Curve fitting can be performed using iterative methods. The best results are obtained with nonlinear least square (NLLS) fitting. This method applied to the Cole model and introduced by MacDonald [26], estimates the best coefficients through an error minimization process. The NLLS method can be applied on the frequency or impedance domains, each having specific advantages and drawbacks.

For instance, in the impedance domain, the NLLS approach is a traditional method to effectively estimate $R_0$, $R_\infty$ and $\alpha$, but poor in the estimation of $\tau$, because this approach does not account for frequency. On the other hand, if frequency is considered, computing time may also increase. The same NLLS approach in the frequency domain has also been studied [27] for the estimation of Cole parameters.
3.2.2. *Estimation of Body Composition Parameters*

Several methods have been developed to estimate body composition parameters being Magnetic Resonance Imaging (MRI) [28, 29] and Computed Tomography (CT) [30], two of the most accurate methods for the estimation of body fat. Dual Energy X-ray Absorptiometry (DEXA) and Under Water Weighing (UWW) are other methods that have also shown reliable measurements and have been considered the gold standard for several studies and for the validation of other methods.

Total Body Water content (TBW) is another relevant parameter that can be calculated from FM and FFM or estimated using different methods. The dilution method is one of the most precise methods and is commonly utilized as a gold standard.

Notwithstanding the accuracy and consistency of these methods, they have the following significant drawbacks:

- Underestimation or overestimation in obese and skinny people, respectively.
- Require expensive equipment and highly qualified technicians to be performed.
- Are usually available only in clinical settings or hospitals.
- Are invasive and require blood samples and administration of tracers or involve exposure to X-ray radiation, thus they cannot be performed at short intervals.
- Are inadvisable for patients with metal or cochlear implants or with cardiac pacemakers.

As an alternative method, BIA has a solid theoretical framework; the estimation of Body Composition by this method overcomes most of the aforementioned shortcomings. However, as a main drawback, BIA still has not gained a full consensus for use in all types of human morphologies and medical conditions. The estimation of these parameters in healthy individuals is satisfactory; however, reliable measurements in individuals with abnormal ICF/TBW ratios have not presented the same consistency. Therefore, the accuracy of bioimpedance methods is dependent upon external factors such as the diversity of the population and the validity and robustness of the electrical model of the analyzed tissues.

First efforts to use bioimpedance to estimate Body Composition parameters initiate with use of two frequencies 1 and 100 kHz, to estimate ECF and TBW based on the Cole model. Further methods have also been developed to estimate TBW from regression equations for measurements from wrist to ankle at a single frequency of 50 kHz and subject height.

More recently, complex and more elaborate methods that require multi-frequency impedance meters were proposed to determine Body Composition Parameters using BIS [31]. Although different formulas and approaches were developed, most of them used Fricke’s electrical model (figure 2.2) and the Cole model to extrapolate values of $R_0$ and $R_\infty$. 

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As a second step, one of the most accepted approaches estimates ECF using a formula derived from De Lorenzo et al. method [32] that is based on Hanai’s mixture conductivity theory [33].

\[
ECF = K_e \left( \frac{H^2 W^{1/2}}{R_e} \right)^{2/3}
\]

Equation (3.2)

In equation 3.2, H and W are the height and weight of the subject, respectively. Obtained from the model fitting, \( R_e \) is the resistance of the extracellular fluid and is approximately equal to \( R_0 \). The dimensionless constant \( K_e \) depends on the shape factor of the human body, which is approximated as the sum of 5 cylinders (limbs and trunk) using one of two approaches.

In the first approach [34], values of \( K_e=0.306 \) and \( K_e=0.316 \) were suggested for men and women, respectively, based on measurements performed with the bromide dilution method.

In a second approach Moissl et al [35] proposed the estimation of \( K_e \) for ECF was from Hanai’s mixture theory and Body Mass Index (BMI). Using the formula shown in equation 3.3, \( K_e \) is determined in terms of BMI, thereby taking into account subject morphology.

\[
K_e = \frac{a}{BMI + b}
\]

Equation (3.3)

The parameters \( a \) and \( b \) of this equation were obtained from regression analysis and found to be \( a=0.188 \) and \( b=0.2883 \). For the determination of ICF, it was first necessary to obtain the resistance \( R_i \) from equation 3.4.

\[
R_{\infty}^{-1} = R_e^{-1} + R_i^{-1}
\]

Equation (3.4)

Then, ICF was estimated from the methods outlined by De Lorenzo et al. or Moissl et al. [32]. In the first method, equation 3.5 is expanded and solved numerically. In this equation, \( K_p \) is the resistivity ratio of ICT to ECF and the values of \( K_p \) are 3.82 and 3.40 for men and women, respectively.

\[
\left( 1 + \frac{ICF}{ECF} \right)^{5/2} = \frac{R_e + R_i}{R_i} \left( 1 + K_p \frac{ICF}{ECF} \right)
\]

Equation (3.5)

Following the second approach, ICF is estimated according to equation 3.6.

\[
ICF = K_i \left( \frac{H^2 W^{1/2}}{R_i} \right)^{2/3}
\]

Equation (3.6)

The constant \( K_i \) is estimated in terms of BMI using equation 3.7 with \( c=5.8758 \) and \( d=0.4194 \), which were determined by cross validation as described in [35].

\[
K_i = \frac{c}{BMI + d}
\]

Equation (3.7)
The determination of TBW can also be calculated with different methods, recent approaches suggested to follow a similar methodology as the one used by De Lorenzo and Moissl to estimate extra- and intra-cellular fluid. Equation 3.8 presents a formula proposed by Jaffrin et al. [36].

\[
TBW = K_t \left( \frac{H^2 W^{1/2}}{R_\infty} \right)^{2/3}
\]

Equation (3.8)

Traditionally, TBW is obtained after the estimation of intra- and extra-cellular volumes using equation 3.9.

\[
TBW = ECF + ICF
\]

Equation (3.9)

Finally, Fat Free Mass (FFM) is estimated with equation 3.10 with a hydration constant of 0.732, which means that 73.2% of the FFM is water.

\[
FFM = TBW / 0.732
\]

Equation (3.10)

Different methods to the equations outlined above have been proposed to determine body composition parameters by means of BIS, but this section focuses on approaches that have gained more consensuses.

3.3. Impedance Cardiography (ICG)

Impedance Cardiography (ICG) is traditionally considered a part of impedance plethysmography and essentially focuses on the determination and analysis of the hemodynamics of the heart. The ICG technique can be performed using a transthoracic or a whole body approach. The first approach, known as Thoracic Electrical Bioimpedance (TEB), is most likely the most studied due to the potential use for diagnosis and treatment of a number of clinical conditions. In this technique, an alternating current (AC) signal between 0.5 mA and 4 mA is injected into the thorax. A single frequency signal in the range of 50 to 100 kHz is commonly applied and measured with a tetra-polar set-up and a lateral spot electrode array as suggested in the literature by Woltjer [37]. A traditional measurement set-up is presented in figure 3.3, where current injecting and voltage sensing electrodes are placed around the abdomen at the level of the sternum and the xiphisternal junction and at the base of the neck.

The inner voltage sensing electrodes, separated by a distance L, define a thoracic volume over which a potential difference is measured and an impedance Z is estimated by means of Ohm’s Law.

The ICG technique has the distinctive capability to provide beat-to-beat cardiovascular information. The blood has twice the conductivity of muscle and several times higher conductivity than other types of tissue [38, 39]. Blood also produces pulse-dependent impedance changes that are easily recorded. Important curves can be obtained from thoracic electrical bioimpedance (TEB) measurements, such as impedance changes (ΔZ) and its first time derivative.
(dZ/dt). Using the correlation found by Lababidi et al [40] between the ICG waveforms and mechanical events of the cardiac muscle, characteristic points from the ECG and ICG signals can be used to estimate clinically important systolic time intervals and hemodynamic indices.

3.3.3. **Sigman Effect**

The singularity of the blood to change resistivity with variations of the blood flow was first observed by Sigman et al. [41]. In this phenomenon called Sigman Effect, blood resistivity decreases when its velocity (flow) increases.

The principle of this phenomenon can be explained by making use of figure 3.4 and knowing that blood contains non-spherical bodies called erythrocytes or red blood cells. As shown in figure 3.4A when the aortic valve is closed, no flow is present and erythrocytes in the aorta artery, adopt a random orientation that makes it difficult for electric current to flow. Less current indicates higher resistance (lower conductivity) in the axial direction. Once the aortic valve is opened, blood flows and aligns erythrocytes parallel to the direction of blood flow (figure 3.4b). Parallel alignment of red blood cells facilitates greater current flow through the artery by producing a lower resistivity.

![Figure 3.3 Traditional tetrapolar measurement set-up for TEB using a lateral spot electrode array.](image1)

![Figure 3.4 Sigman effect presented in the Aorta. A) Aortic valve is closed and no blood flow is experienced. B) Aortic valve is open so blood flow is present.](image2)
3.3.4. TEB Cylinder Model

Traditionally, the principle and characterization of Impedance Cardiography measured through TEB has been explained using the double cylinder model based on theory developed by Nyboer et al. [42, 43].

Under this model the distribution of the blood in the thorax can be determined by measuring the fluctuation of impedance in the sternum. Excluding breathing and the changes in distribution of blood, impedance components of the thorax remain basically constant. The former assumptions allow thoracic impedance measurements to estimate the blood distribution in the thorax on a beat-to-beat basis because it is a function of heart cycle activity.

In this model, presented in figure 3.5a, the volume of the thorax is characterized by a double cylinder. Both cylinders have a length L, representing the distance between voltage sensing electrodes (figure 3.3). In this model the inner cylinder represents the blood compartment (Aorta) and the outer cylinder represents the tissue compartment (muscles, bones, etc.). In the same figure the areas of tissue and blood cylinders are denoted by $A_t$ and $A_b$, respectively.

As electric current passes through this section of the thorax, the total impedance $Z$ is estimated as the parallel impedance of the blood and the tissue. The electric circuit model is shown in figure 3.5b. From the electric model, total impedance ($Z$) can be obtained using equation 3.11:

\[
Z = \frac{Z_tZ_b}{Z_b+Z_t}
\]

Equation (3.11)

Considering a cylinder with uniform cross sectional area $A$, length $L$ and resistivity of the material $\rho$, the resistance $R$ of this volume can be calculated with the equation 3.12:

\[
R = \frac{\rho L}{A}
\]

Equation (3.12)
Because the interest of this work is to study changes in $Z$, equation 3.11 can be differentiated with respect to $Z_b$ and simplified into the following expression:

$$\frac{dZ}{dZ_b} = \frac{d}{dZ_b} \left( \frac{Z_t Z_b}{Z_b + Z_t} \right) = \frac{Z^2}{Z_b^2}$$

Equation (3.13)

If the volume $V_b$ of the inner cylinder, which represents blood, is expressed using equation 3.12 in terms of impedance $Z (A=\rho L / Z)$ and assuming blood resistivity of approximately 150 $\Omega$/cm, the following expression is obtained:

$$V_b = L A_b = \frac{L^2 \rho_b}{Z_b}$$

Equation (3.14)

To get an expression in terms of volume change, equation 3.14 is differentiated with respect to $Z_b$ as:

$$\frac{dV_b}{dZ_b} = -\frac{\rho_b L^2}{Z_b^2}$$

Equation (3.15)

Finally, the equation that relates a change in volume due to a change of impedance is obtained by solving equation 3.13 for $Z_b^2$ and substituting it in equation 3.15

$$dV_b = -\frac{\rho_b L^2}{Z^2} dZ$$

Equation (3.16)

Equation 3.16 is valid (linear) only under the assumption that $dZ << Z$ and the minus sign denotes that an impedance increase correspond to a volume reduction.

3.3.4. Recent Model Electrical Velocimetry

More recently, a new model was proposed by Bernstein et al. [44] with a new way to interpret an ICG signal. In this model, the static transthoracic base impedance ($Z_0$) was estimated from a cylindrical volume of length $L$ and thoracic resistivity $\rho_T$. Equation 3.17 represents the static transthoracic base impedance $Z_0$ with no respiratory or cardiac influence.

$$Z = \frac{U(t)}{i(t)}$$

Equation (3.17)

However, a transthoracic measurement over time also comprises a respiratory and a cardiac component. Thus, when the three components are superimposed into a time-variable transthoracic impedance signal $Z(t)$, the signal can be represented with equation 3.18 as an addition of static ($Z_0$), respiratory ($\Delta Z_R$) and cardiac ($\Delta Z_b$) components.
If the respiratory component $\Delta Z_R$ is suppressed, the signal $Z(t)$ will be equivalent to the parallel of the DC component $Z_0$ and the dynamic AC component $\Delta Z_b(t)$, also known as cardiogenically induced pulsatile impedance change. The new transthoracic impedance signal $Z(t)$ is described by equation 3.19:

$$Z(t) = Z_0 \parallel \Delta Z_b(t) = \frac{1}{Z(t)} = \frac{1}{Z_0} + \frac{1}{\Delta Z_b(t)}$$  \hspace{1cm} \text{Equation (3.19)}$$

Because the static impedance $Z_0$ is comprised of different impedance components, it can be represented as the reciprocal addition of tissue impedance $Z_t$, blood impedance $Z_b$ and extravascular lung water impedance $Z_{LW}$. Equation 3.20 represents the static impedance $Z_0$ and its conformation.

$$Z_0 = Z_t \parallel Z_b \parallel Z_{LW} = \frac{1}{Z_0} = \frac{1}{Z_t} + \frac{1}{Z_b} + \frac{1}{Z_{LW}}$$  \hspace{1cm} \text{Equation (3.20)}$$

The graphical representation of equations 3.19 and 3.20 is depicted in the multi-compartmental parallel conduction model shown in figure 3.6. In this model, the static base impedance $Z_0$ can be interpreted as a triple layer cylinder with constant length $L$. The core cylinder represents the blood impedance $Z_b$ surrounded by a conductive Extravascular Lung Water (EVLW) cylinder with impedance $Z_{LW}$.

The impedance contribution provided by the EVLW tube is only present under a disease condition where an excess of water is experienced. On the top of both inner cylinders, a third tube with non-conductive properties represents the static tissue impedances $Z_t$.

---

**Figure 3.6** Multi-compartmental parallel conduction model of thoracic impedance. $Z(t)$ represents a time variable thoracic impedance estimated from a tetra-polar configuration with an injected current $I$ and voltage sensing $V$. $Z(t)$ comprises a parallel contribution of static base impedance $Z_0$ and time variable and pulse dependent blood component $\Delta Z_b(t)$. The static impedance components for tissue ($Z_t$), blood ($Z_b$) and EVLW ($Z_{LW}$) represent the parallel components of $Z_0$. 

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Finally, it is important to mention that the cardiogenically induced pulsatile impedance component $\Delta Z_b(t)$ has two components. The first one is due to a velocity-induced change in the resistance when blood is flowing in the axial direction due to the Sigman Effect. The second component is a change in the transthoracic resistance related with the stroke volume-induced expansion of the ascending aorta with blood.

3.3.5. **Characteristic Points and Parameters Estimation**

The estimation of relevant parameters is based on the accurate detection of relevant spots called *Characteristic Points*. These points in the ICG and ECG signals represent the electrical and mechanical activity of the heart during a cardiac cycle. Figure 3.7 presents the ICG and ECG signals with localization of the most relevant characteristic points.

With the localization of the characteristic points diverse systolic time intervals and hemodynamic indices can be estimated from these important parameters:

*Left Ventricular Ejection Time (LVET).* LVET is a systolic time interval representing the time elapsed in ms from the opening to the closure of the aortic valve. This timing parameter gives information about how long it takes for the heart to pump blood out of the left ventricle. It is represented as the time from the B-spot to the X-spot in figure 3.7.

*Pre-Ejection Period (PEP).* PEP is a systolic time interval that represents the electrical-mechanical delay that occurs from the onset of depolarization to the beginning of ventricular contraction that is when the aortic valve opens. In figure 3.7, PEP is represented by the time between the onset of the Q wave in the ECG signal (Q-spot) and the B-spot in the first derivative signal.

*R to Z time (R-Z).* R-Z represents the time span between the R-peak in the ECG and the maximal point in the first derivative curve. In figure 3.7, R-Z corresponds to the time span between the R and Z spots. The R-Z time is used in the estimation of the Heather Index [45, 46], considered an index of the cardiac contractility.

*Heart Rate (HR).* Is a measurement of the number of heartbeats per unit of time, usually beat per minute (bpm). HR can be estimated from the ECG signal with the number of R-peaks per unit of time or through the ICG signal, taking as a reference any of the characteristic points detected. HR is an important parameter used by physicians to assist in the diagnosis and tracking of different conditions.
Stroke Volume (SV). SV is a hemodynamic parameter that represents the volume of blood pumped from one ventricle in each heartbeat. SV is one of the most medically relevant hemodynamic indices that can be obtained using ICG [47, 48].

Traditional methods for the estimation of SV based on Nyboer theory have shown low consistency between the impedance waveforms and hemodynamic variables [49], suggesting a lack of robustness in the equation models to estimate SV and a low sensitivity and specificity of the impedance signal acquisition in some medical conditions. The estimations made for this research were performed using the formula presented in equation 3.21. This formula uses the Electrical Velocimetry model and has been shown to be highly consistent [50].

\[
SV = V_{\text{EPT}} \sqrt{\frac{dZ(t)/dt_{\text{max}}}{Z_0}} \cdot \text{LVET}
\]

Equation (3.21)

Equation 3.21 was proposed by Bernstein et al. [44, 51]; \( V_{\text{EPT}} \) represents the volume of electrically participating thoracic tissue (mL), \( dZ(t)/dt_{\text{max}} \) also depicted in figure 3.7 the maximal first time derivative of impedance (ohm s\(^{-2}\)), \( Z_0 \) the measured base transthoracic impedance (ohms) and LVET the left ventricular ejection time (s).
4.1. Introduction for Theory in Textile Electrodes

Most textile products and fabrics are made from primary materials called fibers. The fibers depending on their origin can be classified as natural or manufactured; natural fibers are derived from natural materials and manufactured fibers are created through technology. Table I presents some common examples of natural and manufactured fibers.

<table>
<thead>
<tr>
<th>Natural Fibers</th>
<th>Manufactured Fibers</th>
</tr>
</thead>
<tbody>
<tr>
<td>cotton</td>
<td>nylon</td>
</tr>
<tr>
<td>linen</td>
<td>metallic</td>
</tr>
<tr>
<td>wool</td>
<td>polyester</td>
</tr>
<tr>
<td>silk</td>
<td>spandex</td>
</tr>
<tr>
<td>cashmere</td>
<td>carbon</td>
</tr>
</tbody>
</table>

Yarn is composed of fibers; however, whether a fiber is suitable to create yarns or fabrics depends upon physical, chemical and mechanical properties of the fiber. This research only considers the general properties relevant for the manufacture and good performance of textile
electrodes. However, it is important to consider that the characteristics and the way a textile garment will perform during its useful life will depend on:

- Fiber characteristics
- Type of yarn
- Fabric construction
- Special finishes used

### 4.1.1. Fiber Properties

**Length**
Depending on fiber length, they can be classified as *staple fibers*, commonly measured in centimeters or inches and with relatively short length, or as *filaments*, commonly measured in yards or meters and with indefinite length. The length of the fiber will have an influence on the appearance of the yarn. For instance, filaments commonly produce a flatter yarn surface while staple fibers produce more fiber ends along the surface. The length of the fiber is also related to its strength; shorter staple fibers are weaker than filament fibers.

**Strength**
This property determines the resistance of the fiber to stretching due to forces applied parallel to the fiber axis. Strength is also related to the resistance of the fiber to tearing apart when tension is applied. The force required to break a fiber is known as tensile strength. The strength with respect to linear density is known as *tenacity* and is commonly measured in grams per tex or grams per denier. Some examples of fabrics with strength fibers are Nylon, Polyesters, and Polypropylenes.

**Elongation and Elastic Recovery**
Elongation determines how much a fiber can be stretched under a tensile load without considering the capacity of the fiber to return to its original position. Elastic recovery is the property of the fiber to instantaneously recover its initial position once the stress is removed. Some examples of fabrics with elastic fibers are Spandex and Nylon.

**Electrical Conductivity**
Electrical conductivity is the ability of a fiber to facilitate the transfer of electrical charges. Good conductivity is related to low absorbance of the fiber.

### 4.2. Fabric Construction Methods

Fabrics can be constructed using different methods; in each construction method, yarns are arranged following a specific pattern. Each method requires a specific technique and process to be manufactured. This section will describe only the construction methods part of a general classification.
4.2.1. **Woven Fabrics**

A woven fabric is made of interlacing systems of yarn. In this method the fabric is built by interlacing yarn strands at right angles. The yarns in the lengthwise-direction are known as *warp yarns* while the crosswise yarns are known as *weft yarns*. The most common woven fabric is plain-weave. Woven fabrics are not usually stretchable. However, this also depends on whether the yarns used to build the fabric have elastic properties. Figure 4.1 shows an example of a woven fabric.

4.2.2. **Knitted Fabrics**

A knitted fabric is made by interlocking yarns into loops. The process of producing knitted fabrics can use a single yarn or an array of yarns moving in one direction. When a single yarn is used, the yarn is looped through itself to build a chain or a row of stitches. These rows are connected side by side to produce the fabric. In knitted fabrics the rows of stitches looped in a lengthwise direction of the fabric are called *wales* while the crosswise rows of stitches are known as *courses*. Figure 4.2 shows the interlocked construction of a knitted fabric.
Depending on how the interlocking of the loops is done, knitted fabrics can be classified as *weft knit* or *warp knit*. In *weft knit* fabrics, the yarns are fed in a crosswise direction, perpendicular (right angles) to the length or the direction of growth of the fabric, and the yarn is interlocked across the fabric. In *warp knit* fabrics, the yarns build wales in the lengthwise direction of the fabric.

Knitted fabrics are stretchable; this is an important characteristic in the manufacture of wearable garments because they are more comfortable and fit better to the shape of the body.

*Intarsia Knitting*

Intarsia is a knitting technique which allows the insertions of extra sets of yarns. The extra sets of yarns are interlaced in to the knitted loops. This technique allows the insertion of a different type of yarn (i.e., conductive yarns) in specific zones within the garment.

### 4.2.3. Nonwoven Fabrics

Nonwoven fabrics are textile structures made by interlocking or bonding fibers by means of chemical, thermal, mechanical or solvent techniques. Nonwoven fabrics have recently been increasingly produced and are used in areas like clothing, automotive or healthcare industries.

*Terrycloth Fabric*

Terrycloth is a type of fabric commonly used in the manufacture of towels and is characterized by a thick surface of loops. Terrycloth can be produced using weaving or knitting techniques. There are different types of terry, for instance French Terrycloth only has loops on one side of the fabric and has stretchable properties. The use of loop structures in stretchable wearable garments, such as socks, significantly improves the contact of the fabric against the skin. Figure 4.3 shows a fabric with Terrycloth construction.

![Figure 4.3. Terrycloth construction with a surface of loops.](image)

**NOTE**- The source of information used in sections 4.1 and 4.2 is presented in references [52, 53].
4.3. Electrode Classification

A skin electrode is a contact sensor that creates an electrical interface between the body and the measurement system to allow electrical charges to flow through the tissue and to sense endogenous biopotential.

Electrodes can be classified into two types: polarizable electrodes and non-polarizable electrodes. In the ideal case of perfectly polarizable electrodes, also known as totally polarizable electrodes, no electrode reactions occur when an electrical potential is applied. Therefore, no charge will flow across the electrode-electrolyte interface, and the electrode will behave as a capacitor. Any current flowing through a lead with this type of electrode is a displacement current.

In contrast, in perfectly non-polarizable electrodes the electrode potential will not change from its equilibrium potential when a current density is applied. In this type of electrode, current flows freely across the electrode without producing any potential. An ideal non-polarizable electrode behaves as a resistor with a nominal resistance value that is ideally zero.

Neither perfectly polarizable electrodes nor perfectly non-polarizable electrodes can be manufactured, although similar characteristics can be obtained. For the characteristics previously mentioned, polarizable electrodes are more suitable for sensing biopotentials, while non-polarizable electrodes are more suitable for current stimulation [54].

4.4. Traditional Ag/AgCl Electrodes

Silver-silver chloride (Ag/AgCl) electrodes belong to the category of non-polarizable electrodes; therefore, they allow the transfer of charges and produce very little voltage, ideally none. On the other hand, when used in combination with measurement instrumentation that has high input impedance, ideally infinite, these electrodes are very suitable for measuring electrical biopotential as they are completely polarizable electrodes with extremely low over-potential. Therefore, Ag/AgCl electrodes are the most commonly used electrode for non-invasive physiological measurements.

The reaction between the metal, Ag, and the salt, Ag/AgCl, produces fast electrode kinetics that facilitates current flow through the electrode, improving interfacing between the electrode and skin. This type of electrode has been shown to have several favorable characteristics such as stable behavior, non-toxic composition, low frequency noise and low cost. In figure 4.4, an example of an Ag/AgCl electrode is presented.
The electrodes shown in fig 4.4 use a conductive and sticky gel that improves the electrical properties of the interface between the electrode and the skin and at the same time helps to attach the electrode to the surface of the skin. This electrolytic gel reduces electrode polarization impedance to improve charge transfer.

4.5. **Textile Electrodes**

According to electrode theory, an effective way to evaluate electrode performance is based on its polarization and electrode impedance. Improvement of the conductive properties of the textile material in the electrode will result in good electrode polarization properties and low electrode impedance. Thus, the main problem faced by textile electrodes is the poor conductivity exhibited by the textile material and by the outer layer of the skin.

For this reason, textile electrodes are often made by adding conductive material to the textile. The conductive material can be integrated into the textile as metal fibers (silver, platinum, stainless steel, gold, etc.) in the manufacturing process by coating or laminating the conductive material (conductive polymers) or by embroidering the conductive yarn over the textile structure [55].
Many common electrodes available on the market and used in the biomedical field contain an electrolytic gel that attaches the electrode to the body and improves the skin-electrode interface. Textile electrodes, on the other hand, do not contain an electrolytic gel or fluid. This is an important factor that impedes the ion-electron circulation at the skin-electrode interface, which affects the electrode polarization and electrode impedance. In figure 4.5 two examples of textile electrodes used in this work are presented.

4.6. Skin-electrode Interface

As previously mentioned, the main function of an electrode is to interface between the surface of the skin and the measurement instrumentation, thus converting ionic current from the biological tissue into electronic current and vice versa. In a measurement set-up, the electrode is an element that is sensitive to several factors, which can be affected by several sources of interference. Among these factors, the skin-electrode boundary is most likely the factor with the largest influence on the overall performance of the electrodes.

As explained earlier in this work, the presence of an electrolytic medium, such as a gel, improves the skin-electrode interface. However, in textile electrodes, the lack of electrolytic gel at the skin-electrode boundary produces a deficiency in the conductive properties and a higher sensitivity to external disturbances.

4.7. Skin-Electrode Model

It is important to make a general explanation of the skin conformation. The skin is made of three principal layers: the epidermis, the dermis and a subcutaneous layer. The outer layer, the epidermis, is divided in three sub-layers and is the most significant in the skin-electrode interface. The controlling factor in the epidermal characteristics is the outer sub-layer known as the stratum corneum, which is basically comprised of dead material and can be considered as a membrane that is semi-permeable to ions.

Taking this previous information into account, a circuit model is described in figure 4.6. In this model, the parallel circuit \( R_{\text{skin}} \parallel C_{\text{skin}} \) represents the epidermis, specifically the stratum corneum. This upper skin layer consists of dead cells that have a large contribution to the total impedance. The dermis and the subcutaneous layer have a purely resistive behavior, and they are represented as \( R_{\text{sc}} \) in this model.
The interface between the tissue and the electric conductor is an ionic conductor. The ionic conductor is the electrolytic medium that can be represented by a resistance in series. In the case of dry textile electrodes, the electrolytic medium will be determined by the presence of sweat or by the natural skin humidity of each person. For instance, in a person with dry skin conditions and no presence of sweat, the value of $R_{em}$ will be high.

The sweat produced by the sweat glands is a significant conductive component in the skin-electrode interface and is represented in the circuit by $R_{em}$. The presence of sweat produces more charges between the electrode and the skin; with a higher number of charges, a lower electrode impedance polarization is expected.

In the same model, the parallel circuit $R_{electrode}/C_{electrode}$ represents the electrode surface, which is where the conversion of ionic current to electric current takes place. The contact between a metallic electrode and an electrolyte such as an ionic conductor produces an electrochemical reaction that results in an ion-electron exchange. This ion-electron exchange behaves as a capacitor with a double layer of charges created at the interface.

In the end of the circuit, $U_{SE}$ represents two voltage sources at the skin and electrodes. Consequently, the total impedance expression for this circuit is shown in equation 4.1.

$$Z_{skin-electrode} = R_{SC} + \frac{R_{skin}}{1 + j\omega R_{skin} C_{skin}} + R_{em} + \frac{R_{electrode}}{1 + j\omega R_{electrode} C_{electrode}}$$

Equation (4.1)

For dry electrodes such as textile electrodes, sweat between the skin-electrode interface creates impedance variations.
4.8. Electrode Polarization Impedance in Textile Electrode

The electrode polarization is produced by a displacement of positive and negative electric charges in the skin electrode interface to opposite ends of the electrode. The impedance is obtained using the current (I) flowing through the tissue under study, and the voltage drop (V) is directly affected by the electrode polarization. Conventional Ag/AgCl electrodes contain an electrolytic medium with conductive properties that facilitates charge transfer between the electrode and the skin, thus reducing the value of the polarization impedance $Z_{ep}$; however, in dry-textile electrodes the lack of this electrolytic medium may increase $Z_{ep}$. The large resulting value of $Z_{ep}$ may contribute to amplify the effect of capacitive leakage and cause measurement artifacts; if $Z_{ep}$ is large enough, it completely prevents measurements.

4.9. Textile Characteristics of a Textrode for EBI

The development of a textile electrode requires knowledge of both electrode and textile theory. The right choice of material and fabric construction, with suitable textile properties for the specific application, will result in a better electrode performance.

From the textile point of view, several manufactured characteristics can be chosen to generate an electrode with suitable properties. Among the most significant properties are the material used or type of fiber (cotton, polyamide, spandex, etc.), the properties of the fibers (length, elasticity, strength, etc.), the construction method of the fabric (knitted, woven, non-woven, etc.) and the final treatment (coating) [56]. The appropriate selection of these parameters could result in an electrode with effective skin-electrode interface and good characteristics with respect to the following: conductivity, strength, smoothness, moisture absorption, drapeability, stretchability and water and chemical resistance.
CHAPTER V

TEXTILE GARMENTS FOR EBI MEASUREMENTS

5.1 Textile Bracelet (Initial prototype)

Description: Textile bracelets for wrist and ankle BIS measurements
Width: 2.5 cm.
Length: Adjustable via Velcro fasteners.
Sensor material: Synthetic wrap-knitted textile material with silver fiber as a conductive element. The sensor was manufactured by Clothing+.
Connection: Snap-buttons.

Drawbacks:
- Low content of conductive material in the sensor
- Reduced skin-electrode contact area
- Not optimal skin-electrode contact

Figure 5.1. Textile Bracelet for wrist and ankle
5.2. Textrode Straps

**Description:** Hand-wrist and foot-ankle functional straps for BIS measurements.

**Sensor material:** Shieldex Techniktex P130+B fabric. Silver-plated, two directional stretchability (wrap-weft) and highly conductive knitted fabric manufactured by Statex. Fabric made of 78% polyamide, 22% elastomer and plated with 99% conductive silver with a surface resistivity <2Ω per square.

**Outer fabric:** Synthetic wrap-knitted fabric.

**Intermediate layer:** Foam.

**Closure:** Velcro fasteners.

**Connection:** Snap-buttons.

**Improvements:**

- Increased content of conductive material in the sensor fabric
- Increased skin-electrode contact area
- Improved skin-electrode contact efficiency by means of an intermediate foam layer

**Figure 5.2. Functional straps hand-wrist and foot-ankle**

**Table I Area of the electrodes**

<table>
<thead>
<tr>
<th></th>
<th>Hand-wrist(cm²)</th>
<th>Foot-ankle(cm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Current</td>
<td>97</td>
<td>208</td>
</tr>
<tr>
<td>Voltage</td>
<td>48</td>
<td>160</td>
</tr>
<tr>
<td>Total</td>
<td>145</td>
<td>368</td>
</tr>
</tbody>
</table>
5.3. **Textrode Belts**

**Description:** Neck and chest belts for ICG measurements.

**Belt material:** Neoprene and synthetic wrap-knitted fabric.

**Sensor material:** Velcro loops made of silver-coated fibers.

**Sensor area:** 4 electrodes of 13.5 cm².

**Closure:** Velcro fasteners.

**Connection:** Snap-buttons.

---

**Figure 5.3.** A) Sensor fabric manufactured by Statex and B) transverse view of the functional strap.

**Figure 5.4.** Textile Belts for neck and chest. A) Placement diagram and B) set of belts with the Velcro electrodes highlighted in red.
CHAPTER VI

INFLUENCE OF SKIN-ELECTRODE INTERFACE ON EBI MEASUREMENTS

6.1. Impedance Estimation Error Caused by Capacitive Leakage

The combination of a high electrode polarization impedance \( Z_{ep} \) in the stimulating lead and the existence of leakage capacitances produces an error in the impedance estimation process. To evaluate the influence of the contribution of both \( Z_{ep} \) and parasitic capacitances on impedance estimation, the measurement scenario presented in section 2.7 has been simplified into the electrical equivalent circuit presented in figure 6.1, which models a current leakage from the measurement load. In this model, the parasitic capacitance is represented by a single capacitor.

![Figure 6.1. Electric model representing a bioimpedance measurement using the 4-electrode configuration](image)
C_{par}, in parallel with the measurement load and the electrode polarization impedance, i.e., $Z_{\text{Load}}$ and $Z_{ep}$, respectively.

Analyzing the circuit presented in the Figure 6.1, the following equations are obtained:

$$Z_m = \frac{V_m}{I_0} = \frac{I_m Z_{\text{Load}}}{I_m + I_{cp}}$$

Equation (6.1)

$$V_0 = I_m (2Z_{ep} + Z_{\text{Load}}) = I_{cp}X_{cp}$$

Equation (6.2)

Solving equation 6.2 for $I_m$ yields the following:

$$I_m = \frac{I_{cp}X_{cp}}{2Z_{cp} + Z_{\text{Load}}}$$

Equation (6.3)

Using equations 6.1 and 6.3 and simplifying the following expression for $Z_m$ results in:

$$Z_m = \frac{I_{cp}X_{cp}Z_{\text{Load}}}{2Z_{cp} + Z_{\text{Load}}} = \frac{Z_{\text{Load}}}{1 + \frac{2Z_{cp} + Z_{\text{Load}}}{X_{cp}}}$$

Equation (6.4)

From equation 6.4 it is possible to see that without parasitic capacitance, the measured impedance $Z_m$ would be equal to the impedance of the measurement load $Z_{\text{Load}}$.

### 6.2. Influence of $Z_{ep}$ and $C_{par}$ sensitivity in the measurement

A simple way to analytically study the role of electrode polarization impedance and parasitic capacitance sensitivity in the measurement is to calculate the relative error for different values of $Z_{ep}$ and $C_{par}$. The relative error in the measurement is obtained by first calculating the absolute error using the following expression:

$$\text{AbsError} = |Z_m - Z_{\text{Load}}|$$

Equation (6.5)

where $Z_m$ is the impedance estimated from the electric model in section 6.1, and $Z_{\text{Load}}$ is the impedance of the load. Substituting the expression obtained from equation 6.4 into equation 6.5, equation 6.6 is obtained:
\[
\text{Abs}_{\text{Error}} = \left| Z_m - Z_{\text{Load}} \right| = \left| \frac{Z_{\text{Load}}}{1 + \frac{2Z_{\text{ep}} + Z_{\text{Load}}}{X_{\text{ep}}}} - Z_{\text{Load}} \right|
\]

Equation (6.6)

Simplifying equation 6.6, the error expression is obtained:

\[
\text{Abs}_{\text{Error}} = \left| \frac{Z_{\text{Load}}}{1 + \frac{X_{\text{ep}}}{2Z_{\text{ep}} + Z_{\text{Load}}}} \right|
\]

Equation (6.7)

From equation 6.7, it is easy to observe the role of the parasitic capacitance, i.e., the polarization impedance of the electrode, in the impedance estimation error. Once the absolute error is obtained, the relative error is calculated with equation 6.8.

\[
\text{REL}_{\text{Error}} = \left| \frac{\text{Abs}_{\text{Error}}}{\text{TrueValue}} \right| = \left| \frac{\text{Abs}_{\text{error}}}{Z_m} \right|
\]

Equation (6.8)

Table I presents the values of \(C_{\text{par}}\) and \(Z_{\text{ep}}\) used to simulate the impedance; this simulation is based on the electric model explained in figure 6.1. The results obtained in the simulation are plotted in figures 6.2 and 6.3. The resistance and reactance spectrums were analyzed separately in the frequency range of 3 kHz to 1 MHz.

<table>
<thead>
<tr>
<th>(Z_{\text{ep}} (\Omega))</th>
<th>(C_{\text{par}} (pF))</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>1.5</td>
</tr>
<tr>
<td>100</td>
<td>7.5</td>
</tr>
<tr>
<td>500</td>
<td>20</td>
</tr>
<tr>
<td>1000</td>
<td></td>
</tr>
</tbody>
</table>

Table I: Values of the Electrode Impedance and Parasitic Capacitance

The plot depicted in figure 6.2 shows both the resistance and the reactance spectra; the impedance load is plotted with a solid black trace, while the spectrum of measured impedance, \(Z_m\) (affected by \(C_{\text{par}}\) and \(Z_{\text{ep}}\)), is plotted with colored traces and markers.
Figure 6.3 shows the relative error of the resistance and reactance spectra to better show the influence of $Z_{ep}$ and $C_{par}$. As shown from the analysis of the impedance estimation error and from the results obtained from the simulations, the parasitic capacitance and the impedance electrode polarization clearly influence the impedance value. The spectral plots from figure 6.2 indicate that the error is greater in the reactance component. Although the resistance component is also affected, the relative error is smaller by an order of ten (figure 6.3). From the spectral plots, it is also easy to realize that sources of error, the parasitic capacitance and the electrode polarization impedance, influence the impedance estimation at higher frequencies.

Because parasitic capacitance cannot be predicted or removed completely from a measurement setup, the effect of capacitive leakage increases with the value of the electrode polarization impedance, and the electrode polarization impedance is expected to be larger in dry textile electrodes than in Ag/AgCl electrodes. The error produced in textile-enabled EBI measurements must be sufficiently small so it does not contaminate the EBI spectroscopy data. Therefore, when designing garments for EBI spectroscopy using textile electrodes, the issue of the skin-electrode

Figure 6.2 Impedance spectra with a $C_{par}=7.5\ \text{pF}$ and different values of $Z_{ep}$. A) Resistance spectrum. B) Reactance spectrum.

Figure 6.3. Impedance spectra relative error with $C_{par}=7.5\ \text{pF}$ and different values of $Z_{ep}$. A) Resistance relative error and B) Reactance relative error.
interface must be carefully considered with the aim of achieving low electrode polarization impedance. Natural steps toward the reduction of the electrode polarization impedance would be to increase the surface contact area between the electrode and the skin, to remove dead skin cells from the surface and to keep good contact pressure between the electrode and the skin.

6.3. Estimation of $Z_{ep}$ on EBI

It is expected that textile electrodes have different values of $Z_{ep}$ and hence it is important to quantify $Z_{ep}$. Furthermore, it is expected that $Z_{ep}$ is correlated with the size of the skin-electrode contact area; because the textile electrodes analyzed in this research work have considerably larger area, it is expected to influence the measurement.

To estimate $Z_{ep}$ the following approach was implemented: using feet-to-feet four- and two-electrode configurations, EBI spectroscopy measurements were obtained for the electrodes under study in reference to measurements with a saline tank. The EBI measurements were taken on healthy subjects with the Impedimed spectrometer SFB7 in the frequency range of 3 kHz to 1 MHz. The electrode impedance estimation was assessed against impedance data obtained with the traditional Ag/AgCl electrodes used to analyze the influence of an increasing electrode contact area size.

The obtained results presented in figure 6.4 show that the magnitude of the $Z_{ep}$ for the textile straps is clearly lower than the value estimated for the traditional Ag/AgCl RedDot electrodes over the entire frequency range. A difference of approximately 500 ohms is observed at lower frequencies.
Figure 6.5 shows the magnitude of $Z_{ep}$ estimated for both types of electrodes, the textile and the RedDot. Solid traces show the magnitudes using an initial contact area and dashed traces the magnitude when the area of contact was approximately doubled. $Z_{ep}$ from both electrodes was clearly reduced when the areas were increased and was more evident at lower frequencies.
7.1. Introduction to Home E-Healthcare Systems

Home Health Care is a varied range of health care services provided by professional caregivers in the home of the patient. Home health care offers the advantages of being less costly and more convenient while maintaining strict quality standards and effectiveness demanded with the underlying patient safety aspects taken care of. This type of healthcare service is available to patients who present specific health conditions and will be expanded in the future when health care processes will be further developed. Some of the most common services provided to patients enrolled in these systems include:

- Wound care for surgical wounds or for pressure sores
- Speech therapy
- Monitoring serious conditions with unstable health status
- Caregiver and patient education
- Occupational and physical therapy
- Nutrition or intravenous therapy (IV therapy)
- Diabetic care
- Medication management
- Cancer care

In most of the previously mentioned services, the main difficulty to achieve for caregivers has been to provide home health care in an effective and affordable way as well as taken into account the ethical aspects when technical solutions are to be introduced in the patient's home or in surrounding of relatives. To fulfill these requirements, health care organizational models have moved towards the adoption of enabling technologies including Remote Patient Management (RPM).
Remote Patient Management is a transformative technology founded on an improved organization of care processes that stands on the utilization of a system with the following characteristics:

- Physiological monitoring
- New and specific functions for the caregivers
- Protocol driven decision support
- Information and telecommunication technologies

This technology also relies on a restructuring of the typical business model for care of chronic disease, implementing more active participation of the patients and nonclinical providers, resulting in a reduction of the use of hospitals and nursing facilities and thus a net reduction in total cost [57].

An efficient implementation and utilization of RPM technology into the home health care systems can significantly benefit from the use of textile sensors and garments, especially in the treatment of some chronic and serious conditions such as Heart Failure (HF) and Coronary Heart Disease (CHD). The use of Home Telemonitoring has already exhibited effectiveness in ambulatory patients with HF [58]. Under these conditions, patients with unstable health status require a variety of care services such as daily monitoring, medication management, dietary counseling, patient education, etc. Relevant information about the symptoms and vital signs of the patient can be obtained by effective utilization of textiles and wearable sensors.

For instance, the integration of Electrical Bioimpedance technology into wearable garments can be used to obtain important cardiovascular and body composition information highly relevant in the treatment of heart failure patients. The measurements acquired through patient monitoring on a daily basis using Bioimpedance Spectroscopy (BIS) or Impedance Cardiography (ICG) can be processed to estimate some parameters such as fluid in the lungs (liquid retention), breathing rate, Heart Rate (HR), Pre-Ejection Period (PEP), Left Ventricular Ejection time (LVET) and Cardiac

Figure 7.1 Basic model of a Home Health Care System based on home monitoring of the patient.
Output (CO). The analysis and interpretation of these parameters can help the physicians to provide feedback information to the patient, to adjust the medication on an up-titration phase or as an indicator of the condition of the patient.

Figure 7.1 shows a basic model of a home healthcare system using home monitoring with the essential interacting parts. In this fundamental model, physiological measurements and additional information from the patient sensors are acquired by different sensors and sent to the health professionals using a home gateway. Raw measurements can be stored and processed in medical servers for analysis by health professionals, who can then provide the necessary feedback regarding medical alerts, medication dosage, information and education, etc.

7.2. Motiva System

Philips Motiva is a commercial available TV-based telemmedicine platform developed to provide care management to patients with chronic conditions, e.g., Heart Failure, Chronic Obstructive Pulmonary Disease (COPD) and Diabetes. This home telemonitoring platform is a proven example of a Home Healthcare system that has reached clinical used showing successful cost reduction, patient compliance and enhanced patient survival rates. For instance, reduction in healthcare costs has been evident with use of daily monitoring of vital signs that reduces unnecessary hospitalizations or with the use of a rule-based intelligence and clinical decision support tools in the system. These tools reduce the intervention of caregivers in tasks that do not require high-skill clinical expertise and give them more time for patients for whom their skills can make the most impact.

Patients enrolled in this system are treated following a care plan defined and personalized by caregivers and professionals. The services provided by the system include education, coaching, messages and reminders, and vital sign monitoring. Daily monitoring of vital signs requires patient-caregiver interaction for data interpretation and feedback. The interactivity of the system is based on the use of a television and a secure network connection. Figure 7.2 shows an example of patients and health-professionals interacting with the system. Different type of multimedia content is delivered by TV, such as educational videos, surveys, motivational messages and reminders, feedback about vital signs, messages from the nurse, and modifications to diet or medication intake.

Figure 7.2 The Motiva System showing interactions between patients-system-professionals. Reproduce from [56].
The telemonitoring part of the system consists of a set top box in the home of the patient who allows a secure network connection to the hospital or healthcare center and simultaneous wireless connection to measurement devices. The connection with these devices will enable a daily recording of physiological parameters and vital signs such as weight, blood pressure, blood glucose or blood oxygen if necessary. The Motiva platform is a clear example of a system with the flexibility to treat different chronic conditions and flexibility to tailor treatment of patients. The system is built in a way that can be adapted to reach a broader patient population with other medical conditions and to deliver the same effectiveness and healthcare standards.

7.3. HeartCycle

HeartCycle is a personalized home health care system for the management and treatment of patients with Heart failure (HF), Coronary Heart Disease (CHD) and other related pathologies such as arrhythmia, hypertension and diabetes. The system is being developed as an EU project (FP7 216695) that comprises a group of partners forming the HeartCycle consortium.

The system consists of a double loop as shown in figure 7.3. An inner loop, called patient loop, represents the interaction of the patient with the system. A second outer-loop, or professional loop, represents the participation of the medical professional and the patient in the general process. In the patient loop, the interactivity of the system with the patient will provide support in the daily treatment by presenting changes in patient health and showing their adherence to treatment and its efficacy. In the professional loop, the status of the patient resulting from the interaction with the patient in the patient loop will alert caregivers to adverse events or non-adherence to the care plan.

![HeartCycle system interactivity consisting of a patient and a professional loop. Source [57].](image)
The system relies on an active participation of the patient and the use of vital body signs and other parameters to track health status of the patient. Closer monitoring will also deliver information regarding the effects of medication. In addition, the system provides online education and coaching assistance to the patient in order to create awareness about the impact and the importance of adherence to their treatment to motivate their participation in their own care.

The system will make use of a daily management of the previously mentioned conditions by means of a remote multi-parametric monitoring and analysis of vital signs, symptoms and additional information. Based on information obtained from measured parameters and vital signs, the system will deliver an automated response consisting of the following:

- An event alarm if immediate professional attention is required.
- Modifications on the up-titration medication dosage.
- Therapy recommendations.
- Feedback information regarding health status and the impact of the treatment.

The HeartCycle system uses sensors from previous projects, MyHeart and Sensation, as well as from new developments. The sensors used in the system are grouped in four sensor nodes. The different nodes connected to the HeartCycle Network (HSN) acquire different types of information and they are briefly described below.

- A node with sensors embedded in wearable textiles or connected to a specific body sensor, e.g., IMAGE or BISCUIT.
- A node with sensors from commercial stationary devices, such as weight scales or sphygmomanometers.
- An environmental node that collects parameters of the room environment of the patient such as light, noise level and temperature.
- A home gateway node collecting information from the patient loop.

The sensing technology used in the system will allow the acquisition of several physiological signals that can be processed to extract relevant patient parameters. Some of these parameters are extracted from a set of physiological signals taken from textile and wearable sensors. Some of the textile and wearable sensors used in the system are briefly described below:
**Interactive Multi-Parameter Active Garment Electrodes (IMAGE).** This sensing device, developed by CSEM, consists of a sensor embedded in a shirt to monitor vital signs during exercise. The wearable device includes an electrocardiograph, a bioimpedance meter and a 3D accelerometer. The device is worn on the GEx sleeveless shirt that was developed by Clothing+ and holds the portable device (PDA) to wirelessly acquire measurements from the sensors. The information is processed in the PDA for the extraction of physiological parameters such as respiration rate, cardiac rhythm and activity classification. The parameters are then analyzed and feedback information is provided to the patient during physical activity. Figure 7.4 presents the wearable system consisting of the sleeveless shirt and the PDA device (A) and two active electrodes attached to the shirt (B).

**BioImpedance Monitor for Heart Rate Failure Management (BIM).** The solution consists of a portable bioimpedance meter module (BIM) attached to a textile vest with embedded textile electrodes developed by Clothing+. The BIM module acquires transthoracic bioimpedance measurements, which are wirelessly sent to a portable device such as a PDA. The acquired
measurements are then processed via Bluetooth technology and electrocardiograms, activity levels and respiration rates are extracted. Figure 7.5 shows the portable bioimpedance meter module developed by Philips (A) and the textile vest with its embedded textile electrode developed by Clothing+ (B).

**BioImpedance Spectroscopy Clothing Used Independently aT home (BISCUIT).**
The wearable module is intended for use in patients with chronic heart failure and utilizes BIS and ICG, which can be used to non-invasively measure fluid retention in the lungs, CO and other hemodynamic parameters. Figure 7.6 presents a prototype of the wearable module.

![Figure 7.6 Prototype Vest for the BISCUIT wearable module developed at the Heart Cycle Project. Reproduced from [57].](image)

NOTE- The information presented regarding HeartCycle and Motiva were obtained from the documents cited in references [59-63].
In the initial part of this research, an assessment was made to determine if textile electrodes could perform bioimpedance measurements [paper I]. Using a set of custom-made textile bracelets, wrist-ankle BIS measurements were recorded. Based on the results, two hypotheses were confirmed. The first was that textile electrodes could be used to measure BIS; the second was that when the skin-electrode interface is prepared with conductive gel and textile electrodes slightly moistened, textile electrodes produce very similar measurements to Ag/AgCl electrodes. Figure 8.1 shows minimal variations between Ag/AgCl RedDot electrodes and the moistened textile bracelets, but noticeable differences were present when the skin was not prepared with conductive gel.

Once the feasibility of using textile electrodes for EBI measurements was confirmed, the next step was to find out if a dry skin-electrode interface would affect the performance of textile electrodes.

**Figure 8.1.** Impedance spectra of subject 2 with the Red Dot 3M and textile electrodes and with and without conductive gel preparation [paper I]. Wrist-ankle measurements and resulting A) resistance spectrum and B) reactance spectrum.
For this study [paper II], the performance of two types of textile electrodes was evaluated, our manufactured Textile bracelet used in paper I and commercially available Cuff Electrodes from Textronics, Inc. This paper also analyzed the influence of time on measurements because dry electrodes are likely to improve over time as they are moistened by natural skin perspiration. The results obtained from this study revealed that dry textile electrodes produce repetitive and reliable measurements (Figure 8.2).

Differences in the performance of both types of textile electrodes were more evident in the reactance spectrum; cuff electrodes showed a significantly smaller difference from the reference electrode. This difference was most evident at frequencies below the characteristic frequency. The resistance spectrum, on the other hand, was more similar to the resistance spectrum of the reference electrode. Temporal effects were also more evident in the reactance spectrum. Measurements taken with the bracelet-electrode presented a clear improvement over time in the reactance spectrum, while no improvement was clearly noticed in the cuff electrodes which measurements were already closer to the reference electrode.

This study showed noticeably better performance by the cuff-electrodes, which can be attributed to the conductive material in the sensor of the cuff-electrodes that exhibited higher conductivity and contained a larger conductive surface area compared with bracelet-electrodes.
For the third study [paper III], the influence of the skin-electrode contact area on the estimation of Total Body composition parameters was evaluated. Using a novel approach, tetra-polar EBIS measurements in an electrode ring configuration were performed. The effect of skin-electrode contact area was analyzed by performing measurements with one, two, three or four Ag/AgCl electrodes. The resistance and reactance spectra showed small but noticeable differences when the area increased from one to four electrodes and showed greater differences at higher frequencies.

A similar behavior was observed in the impedance plot and conductance spectrum (figure 8.3), where higher values were obtained with four electrodes at frequencies over 150 kHz in the conductance spectrum. In the impedance plot the difference was noticeable from frequencies below the characteristic frequency.

The estimation of the BC parameters presented an increasing trend from the single to the 4-electrode ring in the case of TBW and ICF. However, ECF was independent of the number of electrodes (figure 8.4). The maximal variation obtained was less than 1.5 liters in the estimation of TBC and ICF and less than 0.19 liters in the estimation of ECF.

![Figure 8.3. Impedance plot (A) and conductance spectrum (B) using an electrode ring configuration increasing the area from one to four electrodes.](image)

![Figure 8.4. Body Composition parameters estimated with one, two, three and four electrodes in a ring configuration](image)
With the knowledge obtained from previous studies, the fourth study [paper IV] aimed to assess the performance of a set of hand-wrist and foot-ankle functional straps in the estimation of Body Composition parameters. The assessment of the functional straps was performed using the measurements taken with Ag/AgCl electrodes in the dual opposite electrode configuration as a reference. This type of configuration consisted of a pair of electrodes placed in diametrically opposite positions of the limb and connected with a cable. The intention of such a specific configuration is to mimic the behavior of the straps that surround the limb, producing more homogenous current density distribution in the current injection area.

The results obtained in this study showed high degrees of similarity in the measurements performed with both types of electrodes. Figure 8.5 shows the resistance and reactance spectra, in both cases the straps exhibited smaller magnitudes. The difference in magnitude is consistent over all frequencies.

The estimation of BC and Cole parameters were slightly different between the strap and the reference electrode. The largest difference obtained in the estimation of the Cole parameter was 2.29% for FFM. Table I presents the mean values estimated with both types of electrodes. The calculated values show an underestimation of $R_0$, $R_\infty$ and FM with the straps but no specific trends with the rest of the parameters.

<table>
<thead>
<tr>
<th>Subject 1</th>
<th>Electrode</th>
<th>$R_0$</th>
<th>$R_\infty$</th>
<th>$f_{\text{char}}$</th>
<th>TBW %</th>
<th>ECF %</th>
<th>ICF %</th>
<th>FM %</th>
<th>FFM %</th>
</tr>
</thead>
<tbody>
<tr>
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<td>333.84</td>
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<td>54.56</td>
<td>41.93</td>
<td>58.07</td>
<td>25.47</td>
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<tr>
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<td>330.05</td>
<td>35.61</td>
<td>55.94</td>
<td>42.12</td>
<td>57.88</td>
<td>23.57</td>
<td>76.43</td>
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<table>
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<tr>
<th>Subject 3</th>
<th>Electrode</th>
<th>$R_0$</th>
<th>$R_\infty$</th>
<th>$f_{\text{char}}$</th>
<th>TBW %</th>
<th>ECF %</th>
<th>ICF %</th>
<th>FM %</th>
<th>FFM %</th>
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<tbody>
<tr>
<td>Ag/AgCl</td>
<td>531.89</td>
<td>341.52</td>
<td>27.64</td>
<td>61.80</td>
<td>41.26</td>
<td>58.74</td>
<td>15.58</td>
<td>84.42</td>
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<tr>
<td>Textrode</td>
<td>505.40</td>
<td>329.83</td>
<td>29.41</td>
<td>62.99</td>
<td>41.89</td>
<td>58.11</td>
<td>13.95</td>
<td>86.05</td>
<td></td>
</tr>
</tbody>
</table>

N.B. The measurements done with the Red Dot Ag/AgCl electrode are dual opposite electrode measurements.
Finally, for the last study [paper V], the knowledge and experiences acquired in the assessment of textile electrodes for BIS were used in the evaluation of a set of custom-made textile belts for the neck and chest. In this study Transthoracic Electrical Bioimpedance measurements were performed using the textrode belts to analyze the feasibility of using the belts for ICG analysis. Measurements were taken in each patient sequentially, i.e., first patients were tested for one minute with the textile belts, immediately followed by a second one-minute test with the reference Ag/AgCl electrodes.

The results showed quality ICG and ECG signals were obtained with the textile belts. No ectopic beats were produced by the belts and R-peak detection was performed efficiently. The detection of characteristic points in the measurements taken by the belts did not present any irregularity because signals did not require any specific signal processing. The same filtering was applied to both sets of signals obtained from the belts and the reference electrodes.

In figure 8.6, the SV estimation had overlapping zones for all patients and a maximal mean value difference of 13.37 ml that was exhibited in subject 2. For Left Ventricular Ejection Time, overlapping zones were obtained in all the patients and a maximum difference in averages between the two electrodes of 39.31 ms was found in subject 2. Similar results were observed in the estimation of the rest of the parameters. Of all the measurements performed with the textile belts, the estimation of the maximum of the first time derivative of impedance (dZ/dt_{max}) had the largest mean variation from 88.4% to 114.8% with respect to the reference electrode (Table II).

<table>
<thead>
<tr>
<th></th>
<th>dZ/dt Inc (Ω/s)</th>
<th>SV (ms)</th>
<th>ET (ms)</th>
<th>R-Z Time (ms)</th>
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<tr>
<td></td>
<td>avg  std</td>
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<td>RedDot 1.03 0.01</td>
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<td>S2</td>
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<td>65.28 10.54</td>
<td>310.19 45.79</td>
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<td>173.14 5.04</td>
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<tr>
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<td>74.83 8.03</td>
<td>360.71 54.33</td>
<td>158.41 12.34</td>
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</table>
CHAPTER IX

GENERAL DISCUSSION & CONCLUSIONS

9.1 Discussion & Conclusions

Results from paper I showed that slightly wet textile electrodes could perform reliable EBI measurements that were as good as traditional Ag/AgCl electrodes. Hence, the remaining work focused on determining the performance of textile electrodes for BIS with a dry skin-electrode interface.

In textile electrodes, the lack of a conductive medium, such as water or electrolytic gel, at the skin-electrode interface increases the sensitivity to disturbances like parasitic capacitances or apparent increments of the electrode polarization impedance $Z_{ep}$. This sensitivity to unwanted disturbances in textile electrodes could be reduced by improving certain characteristics or properties of the textile electrode. The performance of two different types of textile electrodes in paper II revealed that characteristics such as amount of conductive material, type of surface and the size of the skin-electrode contact area did influence these measurements.

Fabrics with sensors made with a high concentration of conductive fibers or material with an exposed or uneven surface, such as the loops in terrycloth, exhibit a better and more efficient skin-electrode contact. In contrast, textiles with sensor structures with especially flat surfaces present a less uniform skin-sensor contact that is more susceptible to motion artifacts.

Paper III showed that estimations of Body Composition parameters changed for electrode-skin contact areas with different sizes. With increases in contact area, smaller values of $R_0$ and $R_\infty$ were obtained that resulted in an overestimation of TBW and ICF. Increases in area are also
expected to reduce the effect of the constriction zone and to reduce what appeared to be an electrode impedance mismatch.

Taking in account the previous reports, the textile straps used in paper IV included the important design and manufacturing modifications: a larger skin-electrode contact area, more conductive fibers in the sensor fabric and a foam layer to improve the skin-electrode contact efficiency. These changes produced noticeable changes in spectra recordings and estimation of BC parameters compared with initial prototypes.

Another modality of EBI studied in section 3.3 of this research work was Impedance Cardiography ICG. In paper V, the feasibility of using a set of neck and chest textile belts for cardiac monitoring was studied. The results showed quality signals for the ICG and ECG curves and reliable estimations of parameters such as SV, LVET, R-to-Z time and HR. The differences observed between the belts and the reference electrodes are most likely attributed to the natural variability of different measurement sessions and not necessarily to the type of electrode. There were no missing beats and no need for special signal processing of the acquired signals from the belts because the same algorithms for processing and detection were used for both types of electrodes. Recently a new development regarding ICG measurements has surfaced in the direction that the origin of the impedance change is not cause by change of volume but by blood flow change [50]. Such development proposes to interrogate the brachial artery on the upper arm to obtain the ICG parameters currently obtained from the neck-to-abdomen measurements. The electrode placement for performing the transbrachial electrical bioimpedance velocimetry measurements, the upper arm, is much more garment-friendly than the neck-abdomen.

The results obtained in this research provide evidence that textile garments with a dry skin-electrode interface, like the ones used here, produce reliable EBI measurements in both modalities: BIS and Impedance Cardiography.

The functional textrode straps produced minor variations in the impedance spectra and negligible differences in the estimation of BCA parameters. Electrode characteristics in the straps like skin-electrode contact area size, amount of conductive material and fabric construction do influence the measurement and have to be considered for the design and manufacturing of the textile electrode and garment. ICG measurements performed with the textrode belts showed consistent measurements, quality signals and reliable estimation of parameters. However, determining if the parameters estimate accurate enough results is also dependent on the clinical application.

The implementation of functional garments for home multi-parametric monitoring in e-Healthcare Systems is not far off and initiatives and projects sponsored by the EU, like HeartCycle, are carrying out important R&D efforts to make this implementation possible.

9.2. Future work

The results achieved show that is feasible to obtain EBI measurements with dry textile electrodes. Now the next step is to carry out a study to validate this method for the estimation of body fluids. In this study, the fluid assessment using a gold standard method such as the dilution method should be used for the validation.
Regarding the design of the garment, an improved prototype that comprises full textile integration with device-electrodes connections and cables embedded in the garment is required.

As previously presented in the discussion, the estimation of SV can be obtained from transbrachial bioimpedance measurements. In this area the impedance signal measured is almost entirely attributed to velocity-induced changes in the blood resistivity. The design and test of an arm-garment for transbrachial electrical bioimpedance velocimetry measurements can be considered for the continuation of this study.

The availability of a textile garment with an integrated EBI spectroscopy device and embedded textile sensors would allow the implementation of EBI-based personalized monitoring health monitoring systems for body fluid distribution and shift current health care policies regarding the monitoring of patients under home-bounded dialysis treatment.
APPENDED PAPERS

**Paper I.** *Textile Electrodes in Electrical Bioimpedance Measurements – A Comparison with Conventional Ag/AgCl Electrodes.*
J. C. Marquez, F. Seoane, E. Välimäki and K. Lindecrantz

**Paper II.** *Comparison of Dry-Textile Electrodes for Electrical Bioimpedance Spectroscopy Measurements.*
J C Márquez, F Seoane, E Välimäki and K Lindecrantz
The XIVth International Conference on Electrical Bioimpedance, Florida USA 2010

**Paper III.** *Skin-Electrode Contact Area in Electrical Bioimpedance Spectroscopy. Influence in Total Body Composition Assessment.*
J. C. Marquez, F. Seoane and K. Lindecrantz
The 33rd Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBC 2011, Boston USA, 2011

**Paper IV.** *Textrode Functional Straps for Bioimpedance Measurements-Experimental Results for Body composition Analysis.*
J. C. Marquez, F. Seoane and K. Lindecrantz
*European Journal of Clinical Nutrition* 2012, Status: Accepted

**Paper V.** *Textrode-enabled Transthoracic Electrical Bioimpedance Measurements. Towards Wearable Applications of Impedance Cardiography.*
J. C. Marquez, M. Rempfler, F. Seoane and K. Lindecrantz
*Journal of Electrical Bioimpedance* 2012, Status: Submitted

NOTE.- All the papers included as a part of this research work were peer reviewed
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*Journal of Electrical Bioimpedance* 2012, Status: Submitted

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Textile Electrodes in Electrical Bioimpedance Measurements – A Comparison with Conventional Ag/AgCl Electrodes.
Abstract— Work has been intensified around the integration of textile and measurement technology for physiological measurements in the last years. As a result nowadays it is possible to find available commercial products for cardiovascular personal healthcare monitoring. Most of the efforts have been focused in the acquisition of EKG for cardiovascular monitoring where textile electrodes have shown satisfactory performance. Electrical Bioimpedance is another type of physiological measurement that can be used for personal healthcare monitoring where the integration and the performance of the textile electrodes has not been investigated that thoroughly.

In this work, the influence of the textile electrodes on the measurements and on the estimation of the Cole ($R_0$, $R_\infty$, $f_C$ and $\alpha$) and body composition (TBW, ICW, ECW and FFM) parameters has been especially addressed. Complex Spectroscopy 4-electrode wrist-to-ankle electrical bioimpedance measurements taken with conventional Ag/AgCl and textile-electrodes on customized bracelets have been compared and analyzed in the frequency range 3 to 500 kHz.

The obtained results suggest that the use of textile electrodes do not influence remarkably on the complex spectral measurements neither in the estimation of Cole nor body composition parameter. In any case any possible effect introduced by the use of textile is smaller than the effect of preparing the skin by the using abrasive conductive paste.

I. INTRODUCTION

PERSUING to reduce costs for society, there is an ongoing shift in paradigm within healthcare towards home healthcare and preventive Personal Healthcare. The emerging home and personal health monitoring applications require a combination of several technologies for the implementation of what is call E-health applications and services.

Textile technology has been identified as the key element to catalyze the proliferation of E-health monitoring application for Home and Personal health care. Thus several research initiatives have been dedicated worldwide to investigate the feasibility of integrating textile technology into physiological Measurements System.

Measurements of Electrical Bioimpedance (EBI) can be used to monitor the cardiovascular personal health care [1, 2] body composition assessment in nutrition [3, 4] and body fluid distribution of patients under peritoneal dialysis [5].

The electrode is one of the most influential elements in an EBI measurement system, because electrodes do not only function as potential sensing elements but also as electrical charge interface between the measurement system and the body. Dry Textile electrodes do not have an electrolyte to facilitate the charge transfer, electrons or ions, from the current injecting leads to the biological tissue and this may influence the EBI measurement.

In this work, the influence of textile electrodes in the acquisition measurements of EBI first and later in the estimation of body composition contents is studied.

II. MATERIAL & METHODS

In this study textile electrodes are compared with conventional Ag/AgCl electrodes with respect to their ability to perform in spectroscopy measurement of complex EBI. The obtained complex spectra have been compared and used for estimation of body composition parameters. The variability of obtained parameters has also been studied.

A. Electrodes & Electrolytic Paste

1) Textile bracelet for wrist and ankle
   - Width: 2.5 cm & length: adjustable, velcro fastener.
   - Inner surface, sensor: Synthetic wrap knitted textile material with silver fibre as a conductive element. Sensor Manufactured by Clothing+ and developed by Elina Välimäki.
   - Application: body monitoring in medical and healthcare applications.

Fig. 1 Textile bracelet electrode prototype for wrist and ankle
• Outer material, garment: knitted cotton with elastane.

2) **Red Dot 3M repositionable monitoring electrodes.**
• Area: 10.1 cm² with a snap-button connector.
• Outer surface: flexible non-woven polypropylene covered with polyethylene film.
• Inner surface: hydro gel conductive adhesive type.
• Application: diagnostic ECG measurements.

3) **EVERY Paste**
• Conductive and abrasive paste manufactured by spes medica.

### B. Measurements & Analysis

Using the 4-Electrode method wrist-to-ankle Electrical Bioimpedance Spectroscopy (EBIS) measurements have been taken in four healthy subjects: 3 male and 1 female. See Fig. 2. The Impedimed SFB7 spectrometer has been used to measure between 3 to 1000 kHz for the four different types of measurement: Red Dot 3M and Textile, with and without paste in both cases. **N.B.** The textile electrodes were slightly wet for the measurements.

![Measurement Protocol](image)

**Fig. 2** Measurement protocol for the performed EBIS measurements with Ag/AgCl and textile electrodes.

### III. RESULTS

#### A. Impedance Spectrum

Figures 3 and 4 show that the spectra of the measurements with Ag/AgCl and with the textile bracelets do not exhibit any marked differences. In Fig. 3 it is possible to appreciate the coincidence on the resistance spectra between the Ag/AgCl and the textile electrode measurements. Fig. 4 shows that the coincidence is high but not as high as in the resistance spectra, especially at low and high frequencies where the spectra of the reactance differ the most.

![Resistance Spectrum](image)

**Fig. 3** Resistance spectrum for wrist-to-ankle measurements of subject 2 with the Red Dot 3M and textile electrodes with and without gel.

![Reactance Spectrum](image)

**Fig. 4** Reactance spectrum for wrist-to-ankle measurements of subject 2 with the Red Dot 3M and textile electrodes with and without gel.

<table>
<thead>
<tr>
<th>R₀ (Ω)</th>
<th>R∞ (Ω)</th>
<th>f_c (kHz)</th>
<th>α</th>
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<tr>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
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<td>3M</td>
<td>Tex</td>
<td>3M_G</td>
<td>Tex_G</td>
</tr>
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<td>595.3</td>
</tr>
<tr>
<td>4</td>
<td>449.4</td>
<td>451.6</td>
<td>465.5</td>
</tr>
</tbody>
</table>

**TABLE I**

**Mean Value of the Estimated Cole Parameters from EBIS Measurements with Textile & Electrolytic Electrodes with & Without Gel**

N.B. the index _G indicates the use of abrasive conductive gel for that type of measurements.
On the other hand, Figures 3 and 4 show that the use of conductive paste introduces more noticeable differences. Fig. 3 shows that resistance increases at all frequencies while Fig. 4 shows an increase in reactance up to frequencies around 150kHz with the use of conductive paste in the case of electrolytic electrodes and from 4 kHz and above of textile electrodes. Notice that above 200 kHz the reactance measured by the Red Dot 3 decreases remarkably.

B. Cole Parameters

Table I contains the mean values for the Cole parameters estimated from each type of measurements and for all the subjects, while Fig. 4 contains the minimum, maximum and mean values for the Cole parameters estimated for subject 4.

1) \( R_0 \) and \( R_\infty \) estimation

The values in Table I indicates that the mean of the observed differences, with the exception of \( R_0 \) estimation for subject 1, are below 5% and occur in both directions.

2) \( \alpha \) estimation

According to Table I, the maximum observed difference in mean of the estimated values is smaller than 2.4%.

3) \( f_C \) estimation

In this case, according to Table I, the mean of the deviations do not reach the 4.4%. In the case of using abrasive electrolytic paste, the use of textile electrodes creates an overestimation of \( f_C \).

4) Influence of application of abrasive conductive paste

Looking at Table I it is possible to see that the differences observed between the estimation done with textile electrodes and with conventional Ag/AgCl is smaller in the case of using abrasive conductive paste than without it for the estimation of \( R_0 \) and \( R_\infty \).

C. Body Composition Analysis

Table II contains the mean values for the Cole parameters estimated from each type of measurements and for all the subjects while Fig. 6 contains the minimum, maximum and mean values for the Body Composition parameters estimated for subject 4.

1) Total Body Water

According to Table II, the use of textile electrodes do not introduce a remarkable difference. In Fig. 6.A) it is possible to observe that the range of the estimations of TBW presents noticeable overlaps. The only remarkable difference occurs for subject 1 where the difference reaches 4%.

2) Body Fluid Distribution

The mean values in Table II for estimation ECF and ICF exhibit differences below 2%, that is similar to the dynamic range of the estimation for a single type of measurement in a single subject, see Fig. 6.B.

3) Fat Mass

In this case the mean values show larger relative differences, up to 14% in subject 1.

### TABLE II

<table>
<thead>
<tr>
<th>Subject</th>
<th>3M</th>
<th>Tx</th>
<th>3M_g</th>
<th>Tx_g</th>
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<th>Tx</th>
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<td>52.8</td>
<td>52.4</td>
<td>51.6</td>
</tr>
</tbody>
</table>

N.B. The ECF and ICF and complementary parameters, related by ECF (%) = 1 - ICF (%) creates an overestimation of \( f_C \).
4) Influence of application of abrasive conductive paste

From Table II, in general no remarkable influence can be observed in the use of textile electrodes with or without conductive paste. But the differences observed in TBW (%) when using textile electrodes are slightly smaller with conductive paste.

IV. DISCUSSION

A. Influence of the use of the Textile Electrodes

Since the textile electrodes do not have any electrolytic properties like an Ag/AgCl electrode, it would be expected that an increase in the Electrode Polarization impedance, Zep, would cause more noticeable changes in the measurements, but such remarkable difference has not been observed. This can be due to the fact the textile electrodes were wet prior the measurement. Wetting the textile electrodes is necessary in order to obtain any measurement and it improves the electrical interface between the electrode and the body. Thus by wetting the surface of the electrode the Zep of the textile electrode might not be much larger than the Zep of the Conventional Ag/AgCl Electrode.

To apply electrolytic gel onto the textile electrode should decrease the Zep as much as using water providing a reliable measurement. The use of electrolytic gel on textile electrodes for EBI measurements has been studied already in [7] reporting good obtained measurement performance.

Motion artifacts are a source of error in any wearable systems and even when the use of knitted structures as pointed in [8] decrease textile structure-related motion artifact, additional tests should be performed to study the influence of activity-related motion artifacts.

B. Influence of the use of the abrasive conductive paste

Preparing the skin using abrasive conductive paste would improve the interface resistance by removing dead cells from the skin, increasing the number of free charges available for the charge transfer process and increasing the humidity of the most superficial skin layer. Reducing the Zep reduces the total impedance of the electrical path that the injecting current should use through the body making the measurement more insensitive to influence of any parasitic capacitance that might be present with the load.

Decreasing the impedance of the current signal pathway would minimize the effect of small increments of Zep that could be introduced by using textile electrodes. There is not much information available about the influence of the Zep on EBI measurements but we see a direct relationship between Zep and the effects of parasitic capacitances that deserves to be investigated further.

The observed slight reduction in the differences estimation R0 and TBW(%) when using textile electrodes is due to the fact the body Composition Analysis method use by the BioImp estimates TBW(%) from R0 and R∞ [9].

V. CONCLUSION

The reported results suggest that the use of these textile electrodes instead of conventional 3M Red Dot does not significantly influence the EBI measurement. The spectral measurements obtained with textile electrodes and with conventional Ag/AgCl agree well and therefore the Cole and the body composition parameter that are estimated from EBI spectra do present any remarkable difference either.

On the other hand the use of abrasive conductive paste does introduce changes in the spectrum that are more noticeable than the changes introduce by the textile electrodes. Therefore any influence that the use of textile might introduce is not as strong as the influence of the use of abrasive conductive past.

This work initially suggests that textile electrodes can be use for EBI measurements for body composition assessment by Bioimpedance Spectroscopy analysis. Nevertheless, the role and influence of Zep on the EBI measurement should be investigated further to extend the work done by Medrano in [10], especially targeting the effects of conductive gels and the relationship of Zep with eventual parasitic capacitances.

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Comparison of Dry-Textile Electrodes for Electrical Bioimpedance Spectroscopy Measurements.
Comparison of Dry-Textile Electrodes for Electrical Bioimpedance Spectroscopy Measurements.

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Abstract: Textile Electrodes have been widely studied for biopotentials recordings, specially for monitoring the cardiac activity. Commercially available applications, such as Adistar T-shirt and Textronics Cardioshirt, have proved a good performance for heart rate monitoring and are available worldwide. Textile technology can also be used for Electrical Bioimpedance Spectroscopy measurements enabling home and personalized health monitoring applications however solid ground research about the measurement performance of the electrodes must be done prior to the development of any textile-enabled EBI application.

In this work a comparison of the measurement performance of two different types of dry-textile electrodes and manufacturers has been performed against standardized RedDot 3M Ag/AgCl electrolytic electrodes. 4-Electrode, whole body, Ankle-to-Wrist EBI measurements have been taken with the Impedimed spectrometer SFB7 from healthy subjects in the frequency range of 3kHz to 500kHz. Measurements have been taken with dry electrodes at different times to study the influence of the interaction skin-electrode interface on the EBI measurements.

The analysis of the obtained complex EBI spectra shows that the measurements performed with textile electrodes produce constant and reliable EBI spectra. Certain deviation can be observed at higher frequencies and the measurements obtained with Textronics and Ag/AgCl electrodes present a better resemblance.

Textile technology, if successfully integrated it, may enable the performance of EBI measurements in new scenarios allowing the rising of novel wearable monitoring applications for home and personal care as well as car safety.

1. Introduction

Textile technology might play an essential role enabling home and personal healthcare systems and applications. Measurements of Electrical Bioimpedance (EBI) are being uses for cardiovascular monitoring[1], body composition assessment [2] and other monitoring applications that could benefit significantly from smart textile technology. The reliability of physiological measurements obtained
with textile-enabled measurements systems must be ensured prior obtaining any satisfactory optimal integration of textile technology in a health-related measurement system. Recent developments in textile technology have made available textile electrodes [3] that can function as sensors in non-invasive physiological measurements[4]. In an EBI measurement the electrode has a determinant role in the system due to its double function in the system: as potential sensing elements and as electrical charge interface between the measurement system and the body. The absence of an electrolyte compound in the composition of textile electrodes may impede the charge transfer from the current injecting terminals into the body, thus affecting the measurement.

In this work the performance of two different kinds of dry-textile electrodes from two manufacturers are compared against the performance of typical Ag/AgCl electrolytic electrode, observing the possible influence of time in the obtained EBI complex spectra.

2. Material & Methods
In this study ankle-to-wrist EBI spectroscopy measurements have been done on three healthy subjects lying supine in a resting state at different times, using two types of textile electrodes and conventional electrolytic electrodes. The EBI spectra obtained with textile and electrolytic electrodes have been studied in order to assess their resemblance.

2.1. Electrodes
For this comparative, three different types of electrodes have been used: textile bracelets, wrist cuffs both shown in Figure 1 A) and B) respectively and sticky electrolytic Ag/AgCl pads.

2.1.1. Textile bracelet. - The bracelet has been custom made with a width of 2.5 cm and adjustable length. The material of the inner surface is the electrode sensor manufactured by Clothing+ with synthetic wrap knitted textile material with silver fibre as a conductive element. The outer material of the bracelet is knitted cotton with spandex and Velcro for fastening it around the wrist and ankle.

2.1.2. Wrist/Ankle Cuffs Electrode. - The cuffs are manufactured by Textronics Inc. and made of polyamide (nylon) 15%, conductive fibres 30%, Spandex 20% and polypropylene 35%. The conductive textile material is knitted in the inner surface of the cuff electrode.

2.1.3. Electrolytic electrode. - The electrode is a repositionable electrode of the RedDot series manufactured by 3M with conductive and adhesive hydro gel. The electrode patch is rectangular with an area of 10.1cm² a snap-button connector.

2.2. Measurements & Analysis
EBI measurements with textile electrodes have been taken at four different times in a time window of 12 minutes with the Impedimed SFB7 spectrometer in the frequency range 3-500 kHz. The obtained EBI measurements have been compared with a final measurement taken with electrolytic electrodes. The comparison, based in a relative error analysis, has been made studying the deviation observed from the complex spectral values obtained with the electrolytic electrodes.

Figure 1. Textile electrodes. Textile bracelet in A) and Wrist cuffs in B).
Figure 2. Spectra obtained from the EBI measurements with both textile electrodes at different times. The spectrum obtained with the electrolytic electrode is shown as a reference value.

3. Results

3.1. Complex EBI Spectrum

Figure 1 contains the EBI spectra obtained with both textiles, dotted trace for the bracelet and continuous trace for the cuff, at two different times indicated by a circular marker. The EBI spectra obtained with electrolytic electrodes is included for comparison purposes with dashed trace. In the case of the resistance spectrum, depicted in Figure 2.a), the only spectrum that is remarkably different from the rest is the one obtained with the bracelet electrodes for t=0. The same EBI measurement taking 12 minutes later exhibit a better resemblance. In the case of the reactance spectrum, depicted in Figure 2.b), only the measurements obtained with the elastic cuff electrodes produce an EBI spectra that closely resembles the spectrum obtained with the electrolytic electrodes.

While Figure 1 contains the obtained spectra from the measurements on a single subject, Figure 2 presents the averaged relative error of complex impedance spectra. The error has been calculated considering the spectra obtained with the electrolytic electrode as reference.

It is clearly observed in Figure 2, that the deviation from the reference value obtained with the cuff electrodes is much smaller than the deviation obtained with the bracelet electrodes. The deviation obtained with the cuff electrodes exhibit also a smaller frequency dependency.

In Figure 3, it is also possible to observe certain time dependency. The relative error obtained with the bracelets exhibit a denoted time dependency that suggests that the error decreases with time. In the

Figure 3. Mean relative error of the resistance and reactance averaged among three subjects for both textile electrodes at different times
The mean impedance relative error depicted in Figure 3 is obtained as the mean of the relative errors for the reactance and the resistance, averaged for the three subjects. Therefore, the differences between resistance and reactance for each of the subjects cannot be evaluated from Figure 3 alone. In order to study the error obtained in the resistance and the reactance as well as the time dependency for each of the subjects, the averaged sum of the relative errors have been calculated for both resistance and reactance spectra specifically at four different times for both textile electrodes. The obtained values are presented in Table 1.

In Table 1, the averaged sum values of the resistance and reactance relative errors are presented. From this table is possible to observe that in terms of the averaged sum of relative error the deviation obtained from the resistance spectrum is smaller than the produced in the reactance. The deviation obtained from de resistance measurements with both type of textile electrodes follow the same general trend seen in Figure 1, time reduces the obtained deviation. Therefore the averaged sum of the relative error for the measurements taken after minute 12 are smaller than the ones obtained at t=0. In the case of the reactance spectrum, the bracelet electrodes follow the general trend previously indicated while the deviation obtained with the cuff electrodes does not exhibit any specific trend.

When comparing the deviation obtained with both type of electrodes, it is easy to notice that for all cases with the exception of one, the deviation obtained with the cuff electrodes is smaller.

4. Discussion
The results obtained showed that the EBI measurements done with both textiles electrodes present reliable and repetitive measurements for the resistance spectrum. The error obtained with the bracelet electrodes for the reactance spectrum is very high while the reactance spectra obtained with the cuffs present exhibit a relatively low deviation.

Certain improvements can be observed on the performed spectroscopy measurements. In the case of the measurements taken with the bracelets an improvement can be observed especially in the reactance spectrum, where the initial error is significantly high. In the case of the measurements obtained with the electrode cuff such improvement cannot be notice, but this can be due to the fact that the error is much smaller. Such a smaller error can be due to the fact that the textile sensor used in the cuff exhibits a higher conductivity and the surface of the conductive sensor is also much larger than in the case of the bracelets. Such higher conductivity combined with a larger surface may improve significantly the electrode-skin interface.

5. Conclusion
The results obtained suggest a general good performance of textile electrodes when measuring resistance. The reactance spectrum obtained with the bracelets is not reliable enough while the reactance spectrum obtained with the cuff electrodes is more reliable. Moreover to be able to assess on their reliability for EBI applications for body composition further analysis must be done, especially

<table>
<thead>
<tr>
<th>RESISTANCE SPECTRUM</th>
<th>REACTANCE SPECTRUM</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject 1</td>
<td>Subject 2</td>
</tr>
<tr>
<td>Bracelet</td>
<td>Cuffs</td>
</tr>
<tr>
<td>0.022</td>
<td>0.056</td>
</tr>
<tr>
<td>0.019</td>
<td>0.058</td>
</tr>
<tr>
<td>0.015</td>
<td>0.056</td>
</tr>
<tr>
<td>0.022</td>
<td>0.054</td>
</tr>
</tbody>
</table>

| 0.020 | 0.056 | 0.120 | 0.080 | 0.043 | 0.016 | Mean | 0.234 | 0.100 | 4.329 | 0.153 | 0.408 | 0.087 |
Regarding the estimation of the Cole parameters like in [5]. With Home monitoring applications in mind, further improvements in the electrode ergonomics and textile design should be done aiming to increase their usability.

The cuff electrodes apparently exhibit a better performance and stability in time and frequency than the bracelet electrodes. As this study gives no clue to whether the difference is due to material or design of the electrode further studies may point out ways to improve the performance of textile electrodes for recording of Electrical Bioimpedance.

In any case the resistance spectra from the EBI measurements obtained with the textile and electrolytic electrodes exhibit a close resemblance. Recent work done on Cole parameters estimation from EBI measurements suggest that EBI applications using Cole model-based analysis can be implemented by only measuring the spectrum of the reactance [6]. Therefore if the latter it is confirmed the implementation Textile-enabled EBI application is not so far ahead.

References

Skin-Electrode Contact Area in Electrical Bioimpedance Spectroscopy. Influence in Total Body Composition Assessment.
Skin-Electrode Contact Area in Electrical Bioimpedance Spectroscopy. Influence in Total Body Composition Assessment.

J. C. Marquez, Student Member, IEEE, F. Seoane, Member, IEEE and K. Lindecrantz, Member, IEEE

Abstract — Electrical Bioimpedance Spectroscopy (EBIS) has been widely used for assessment of total body composition and fluid distribution. (EBIS) measurements are commonly performed with electrolytic electrodes placed on the wrist and the ankle with a rather small skin–electrode contact area. The use of textile garments for EBI requires the integration of Textrodes with a larger contact area surrounding the limbs in order to compensate the absence of electrolytic medium commonly present in traditional Ag/AgCl gel electrodes. Recently it has been shown that mismatch between the measurements electrodes might cause alterations on the EBIS measurements. When performing EBIS measurements with Textrodes certain differences have been observed, especially at high frequencies, respect the same EBIS measurements using Ag/AgCl electrodes. In this work the influence of increasing the skin-electrode area on the estimation of body composition parameters has been studied performing experimental EBIS measurement. The results indicate that an increment on the area of the skin-electrode interface produces noticeable changes in the bioimpedance spectra as well as in the body composition parameters. Moreover, the area increment showed also an apparent reduction of electrode impedance mismatch effects. This influence must be taken into consideration when designing and testing textile-enable EBIS measurement systems.

I. INTRODUCTION

Electrical Bioimpedance (EBI) has shown to be a useful tool for the supervision of the body hydration and fluid accumulation in several applications like Body Composition Assessment (BCA), detection of lymphedema, [1, 2] and also for assessment of cardiac function via impedance cardiography [2]. Traditionally Electrical Bioimpedance Spectroscopy (EBIS) measurements for BCA are performed with electrolytic electrodes in a tetra-polar configuration. Usually, the electrodes are placed on the foot-ankle and hand-wrist to perform measurements for whole body measurements [3]. The electrode produces a skin-electrode interface with an area defined by its size, around 10 cm².[2]

Variations in the quality of the skin electrode interface may affect the electrode polarization impedance (Zep) of one or more electrodes. This unbalance of the Zep can create an electrode mismatch and as it has been reported by Bogónez-Franco [4] affecting the measurement and producing changes in the impedance spectra, sometimes quite markedly. In addition to this, the relatively small size of the contact area may contribute to an irregular distribution of the current that is injected in the limb. This results in the formation of regions with higher current density near the injecting electrode, known as constriction zone [5]. This constriction zone may influence the potential sensed by the neighboring voltage electrodes and consequently influencing on the EBI measurement.

When performing EBIS measurements with Textrodes (Textile Electrodes), it has been observed that even though the spectral data obtained with electrolytic electrodes and Textrodes are very similar, it is possible to observe certain differences especially at higher frequencies and more evident in the reactance spectrum.

In BCA, these alterations on the impedance spectroscopy measurement might lead to erroneous estimations of the parameters describing the body fluid distribution ICF (Intracellular Fluid), ECF (Extracellular Fluid) and TBW (Total Body Water).

Differences observed in the impedance spectra obtained with Textrodes might be related to the increased area of the skin-electrode around the limbs. Under this hypothesis, this work analyses the influence of increasing the skin-electrode contact area on the acquisition of EBIS measurements and the estimation of the BCA parameters.

II. MATERIAL & METHODS

A. Measurement & Protocol

A Tetra-polar configuration Total Right Side for EBIS measurement was used on three healthy subjects lying in a supine position. The measurements were done using an electrode ring configuration and following the protocol shown in Fig 1.

A set of 30 measurements with each 1-4 electrodes was performed using the Impedimed SFB7 spectrometer in the frequency range of 3 to 1000 kHz.

B. Electrode-Ring Measurements

Electrodes were placed in ring configuration around the limbs as shown in Fig. 2. In this way EBIS measurements were performed with one, two, three and four electrodes.
placed around the limbs (hand-wrist and foot-ankle) connected with cables for the current injection and voltage sensing points.

The electrodes were placed at opposite and equidistant positions over surface of the limb following a ring pattern. The electrodes used were the traditional Ag/AgCl RedDot 3M repositionable monitoring electrodes with a conductive area of 10.1 cm².

C. Body Composition & Spectral Analysis

The BCA parameters, TBW, ICF and ECF, were estimated with the BioImp software Multi-Frequency Analysis version 5.3.1.1. [6] and then the mean and the standard deviation values were calculated and compared. The 30 EBI measurements performed were also processed and averaged with Matlab to obtain the spectral and impedance plots and to make a spectral comparison.

III. RESULTS

A. Complex EBI Spectrum

In Fig. 3A the resistance spectrum shows negligible differences at frequencies under 400 kHz, above that frequency the resistance spectrum obtained with four electrodes continuously decreases with frequency while the spectrum obtained with a single electrode decreases up to 700 kHz but it remains constant above that frequency.

The reactance spectrum shown in Fig. 3B presents noticeable differences above 150 kHz. Note that above 800 kHz the reactance obtained with a single electrode changes from a capacitive value to an inductive.

In the impedance plot in Fig. 4A) the differences between 1 and 4 electrodes measurements are noticeable already from frequencies lower than the characteristic frequency i.e. the frequency of maximum reactance.

The conductance plot shown in Fig.4.B) shows a similar behavior than the resistance spectrum with the difference that in the conductance plot, the spectra deviates from each other from 150 kHz and above.

In Fig.5 the estimation of the BCA parameters performed with the four electrode configurations is depicted. In Fig. 5A and Fig. 5C the TBW and ICF parameters present an increasing trend from the single- to the 4-electrode configuration. The maximum changes observed on the estimation of the TBW and ICF is less than 1.5 liters in TBW and ICF. The ECF parameter did not present any particular trend and the maximal variation obtained was less than 0.19 liters.
Table I presents the common behavior observed on the measurements, and it also confirms the trend, with increasing number of electrodes, regarding the negligible differences observed in the estimated value of the ECF parameter and the remarkable difference observed on the values of TBW and ICF, especially the later.

IV. DISCUSSION

A. Skin-Electrode Impedance Mismatch

In EBI measurements the presence of electrode impedance mismatch can affect the impedance estimation as shown in [4]. In this study the obtained EBI spectra agree with the EBI spectra reported in [4] and obtained with electrode mismatch. The deviation observed at high frequencies indicates a kind-of inductive behavior present in the measurements obtained with a single electrode. Since the same effect was observed in the experiments reported in [4], it is very likely that the inductive effect is related with the presence of skin-electrode impedance mismatch. The inductive character seems to decrease when the area of the skin-electrode interface increases. Increasing the area decreases the probability that a mismatch in the skin-electrode impedance will occur.

B. Constriction Zone Size

The size of the constriction zone is also an element that might affect the impedance value when performing an EBI measurement [7]. In the single-electrode measurement the current density in the constriction zone is high and not uniformly distributed in the area neighboring the injecting electrode. On the other hand, in the 4 electrode the injected current will use more interface area and will spread out easier producing a uniform current distribution closer to the injection electrodes.

C. Influence on BCA Parameters

The presence of electrode mismatch or changes in the size of the electrodes area seems to affect the estimation of the BCA parameters. From the deviations observed on the spectral plots at high frequencies, the increment in the values of TBW and ICF was expected. The reason behind this is that the ICF parameter is estimated from the impedance value at infinite frequency, also known as $R_\infty$. Similarly, since the values for the impedance at low frequencies remains practically identical, differences in the ECF estimation are remarkable small.

D. Skin-Electrode increased area

As previously presented in [8], a change on the constriction zone caused by the increased area of the skin-electrode contact interface obtained with textrodes straps, influences on the EBIS measurement Those measurements presented a lower value of the resistance, and consequently estimating lower values for $R_0$ and $R_\infty$.

<table>
<thead>
<tr>
<th>Subject</th>
<th>TBW</th>
<th>ECF</th>
<th>ICF</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>-2.199</td>
<td>0.236</td>
<td>-4.176</td>
</tr>
<tr>
<td>2</td>
<td>-2.098</td>
<td>0.234</td>
<td>-3.801</td>
</tr>
<tr>
<td>3</td>
<td>-1.017</td>
<td>0.860</td>
<td>-2.613</td>
</tr>
</tbody>
</table>

N.B. The mean relative percent difference shown is between the 1 and 4-electrode configurations.
In a similar manner using the 4-electrodes ring configuration around the limb, which increases the area of the skin-electrode interface showed a very similar behavior than when using textrode straps around the limb. In both cases, as shown in 3.A) and 3.B) respectively, the obtained resistance spectra were slightly smaller and the reactance slightly larger at high frequencies, than the spectra obtained using a single Ag/AgCl electrode. It is remarkable to observe that the slope of the reactance spectrum at high frequencies presented in 3.B) for the 4-electrodes ring configuration is smaller than for the reactance spectrum obtained with a single Ag/AgCl. Such behavior agrees with the reactance spectrum obtained with textrode straps reported in [8].

V. CONCLUSION

Changes in the skin-electrode area size influence the acquired EBI spectra, both on the real and imaginary part. The estimation process used to calculate the BCA parameters is clearly influenced, overestimating the volumes of ICF and consequently TBW. These results should be taken into account when studying and comparing the measurement performance of textrode garments.

REFERENCES

Textrode Functional Straps for Bioimpedance Measurements-
Experimental Results for Body composition Analysis.
ORIGINAL ARTICLE
Textrode functional straps for bioimpedance measurements-experimental results for body composition analysis

JC Márquez, F Seoane and K Lindecrantz

BACKGROUND/OBJECTIVES: Functional garments for physiological sensing purposes have been used in several disciplines, that is, sports, firefighting, military and medicine. In most of the cases, textile electrodes (textrodes) embedded in the garment are used to monitor vital signs and other physiological measurements. Electrical bioimpedance (EBI) is a non-invasive and effective technology that can be used for the detection and supervision of different health conditions. EBI technology could make use of the advantages of garment integration; however, a successful implementation of EBI technology depends on the good performance of textrodes. The main drawback of textrodes is a deficient skin–electrode interface that produces a high degree of sensitivity to signal disturbances. This sensitivity can be reduced with a suitable selection of the electrode material and an intelligent and ergonomic garment design that ensures an effective skin–electrode contact area.

SUBJECTS/METHODS: In this work, textrode functional straps for total right side EBI measurements for body composition are presented, and its measurement performance is compared against the use of Ag/AgCl electrodes. Shieldex sensor fabric and a tetra-polar electrode configuration using the ImpediMed spectrometer SFB7 in the frequency range of 3–500 kHz were used to obtain and analyse the impedance spectra and Cole and body composition parameters.

RESULTS: The results obtained show stable and reliable measurements; the slight differences obtained with the functional garment do not significantly affect the computation of Cole and body composition parameters.

CONCLUSIONS: The use of a larger sensor area, a high conductive material and an appropriate design can compensate, to some degree, for the charge transfer deficiency of the skin–electrode interface.


Keywords: textile sensors; textrodes; body composition; textile electrodes; bioimpedance spectroscopy; wrist-to-ankle measurement

INTRODUCTION
Technological improvements in home and preventive personal health care are central issues when trying to modernise health services and reduce costs. New applications require a combination and a clever implementation of several technologies. Textile technology has been identified as a central component in the successful integration of E-health monitoring applications, and the development of functional garments capable of monitoring different physiological parameters will most likely prove to be beneficial.

Therefore, research and efforts related to functional garments have aimed at a feasible integration of textile technology into physiological measurement systems and to quantity the precision they can deliver. Wearable solutions with textile sensors embedded in garments are also gaining market interest. Most of these solutions are focused on monitoring electrical activity of the heart in sport and health applications.

Electrical bioimpedance (EBI), and more specifically bioimpedance spectroscopy (BIS), is another type of physiological measurement in which textile technology could be used. If textile electrodes (textrodes) give reliable measurements, the integration of textrodes into EBI/BIS systems' functional garments will enable novel monitoring applications for a number of health conditions, such as pulmonary function, by means of impedance plethysmography, body fluid distribution on patients under peritoneal dialysis and body composition assessment in nutrition.

In an EBI/BIS system, the electrodes have an essential role for injection of current and for sensing of the resulting voltage; therefore, the performance of the electrodes is essential for the quality of the measurement. In addition, as has been shown earlier, textrodes present a higher sensitivity to signal disturbances compared with traditional electrodes.

The conductive properties of the textile material and the outer layer of the skin, the absence of electrolyte in textrodes, the amount of conductive material, the size of the sensor, effectiveness in the skin–electrode contact and the characteristics of the textile structures are some of the factors that determine the performance of the system and the reliability of the measurements.

In this study, a set of functional textile straps (hand-wrist and foot-ankle) for total right side EBI measurements, with improved characteristics with respect to factors affecting the quality of the measurement, was developed and tested for body composition analysis (BCA). The new straps have higher-than-before content of conductive material in the electrode sensor, an extra material layer to guarantee a better and more effective skin–electrode contact and a more ergonomic design. The impedance spectra and the Cole and total body parameters were then analysed and compared against measurements with standard Ag/AgCl electrodes.
MATERIALS AND METHODS

Measurement and analysis

For this study, EBI spectroscopy BCA was performed on three subjects. By using the set of functional textrode straps forming a tetra-polar electrode configuration, sets of 100 total right side EBI measurements were taken for each subject using an ImpediMed SFB7 spectrometer (San Diego, CA, USA).

The measurements acquired with the textrode straps and the Ag/AgCl electrodes were analysed with respect to differences in impedance spectra and the Cole and the BCA parameters in the frequency range of 3–500 kHz.

Cole and BCA parameters estimation. The Cole and BCA parameters were estimated using the Bioimp software (ImpediMed). The software calculates the Cole parameters using the BIS measurements, and from the Cole parameters ($R_0$ and $R_{\infty}$), the intracellular and extracellular fluid (ICF and ECF) contents are estimated using Van Loan’s method and De Lorenzo formula, both based on Hanai’s theory.18,19 According to these formulas, the values of ECF and ICF depend on the values of $R_0$ and $R_{\infty}$. The rest of the parameters, body water, fat-free mass and fat mass, are estimated from the values of ECF and ICF.

Dual opposite measurements. A dual opposite electrode configuration as shown in Figure 1 was used to record the measurements with the Ag/AgCl electrodes. The configuration consists of two pairs of electrodes, one for current injection and one for voltage sensing. In each electrode pair, the electrodes are placed in diametrically opposite positions of the limb and connected with a cable. Such specific electrode configuration is used to obtain a more homogenous current density distribution in the region of the current injection.

Measurement subjects

The measurements were taken for three healthy male subjects lying supine in a resting state. The physical information of the subjects used to estimate the body composition parameters is presented in Table 1.

Functional garment and electrode

Hand-wrist and foot-ankle functional straps

Functional strap design: The inner and outer surfaces of the textrode straps are shown in Figure 2. In this figure, the sensor zones and Velcro fasteners are presented in the Figure 2b sketch.

Figure 1. Dual opposite electrode connection on the foot-ankle. Two electrodes are used for current injection and two more for voltage sensing.

Figure 2. Inner and outer surfaces of the functional straps. (a) Foot-ankle and (b) hand-wrist straps.

Table 1. Subjects information

<table>
<thead>
<tr>
<th>Height (cm)</th>
<th>Weight (kg)</th>
<th>Age</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject 1</td>
<td>174</td>
<td>96.0</td>
</tr>
<tr>
<td>Subject 2</td>
<td>175</td>
<td>95.8</td>
</tr>
<tr>
<td>Subject 3</td>
<td>182</td>
<td>87.0</td>
</tr>
</tbody>
</table>

All the three subjects were of male gender.
manufactured by Statex (Bremen, Germany). This fabric shown in Figure 4a is made of 78% polyamide, 22% elastomer and plated with 99% conductive silver, which provides a surface resistivity $< 2 \Omega$ per square. The approximate electrode area for current and voltage electrodes for both straps is described in Table 2. A transverse view with the different layers of the strap can be observed in Figure 4b.

Ag/AgCl electrode. The Ag/AgCl electrodes were Red Dot repositionable electrodes manufactured by 3M (St Paul, MN, USA). They are made of non-woven polypropylene, polyethylene, inner surface of hydroconductive and adhesive gel and snap-button connector. The sensor area of the electrode is $10.1 \text{ cm}^2$.

**RESULTS**

**EBI spectra**

The complex impedance spectra obtained with the EBI measurements from the three subjects were very similar and, in general, small differences were observed between the textrodes and the Ag/AgCl electrodes. Figure 5 shows the resistance and reactance spectra obtained for subject 2. In this figure, the mean of 100 spectroscopy measurements obtained with the textrodes and the Ag/AgCl electrodes are presented and compared in blue and red trace, respectively.

The main difference observed in both spectrums is the smaller magnitude exhibited by the textrode functional straps over the entire frequency range. The remaining subjects presented in general a similar spectral behavior.

The s.d. of the spectra for subject 2 is presented in Figure 6. As it can be observed, the magnitude of the data dispersion estimated for each electrode over the whole frequency range is negligible compared with the difference between the two electrode types. For instance in Figure 6a, the maximum STD variations in the resistance spectrum for the textile strap and the Ag/AgCl electrode are in the order of 0.01 and 0.2 $\Omega$, respectively. On the other hand, the maximum difference observed between the two types of electrodes in the same figure is in the order of 0.6 $\Omega$.

In addition, from Figure 6a, it is possible to observe a larger dispersion in the measurements obtained with the texttrode functional straps than with the Ag/AgCl electrodes. A different behaviour is presented in Figure 6b, where the reactance spectrum of the Ag/AgCl exhibits a larger dispersion at high frequencies.

**Table 2.** Area of the electrodes

<table>
<thead>
<tr>
<th>Current</th>
<th>Hand-wrist (cm²)</th>
<th>Foot-ankle (cm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>97</td>
<td>208</td>
<td></td>
</tr>
<tr>
<td>48</td>
<td>160</td>
<td></td>
</tr>
<tr>
<td>145</td>
<td>368</td>
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</tr>
</tbody>
</table>

**Figure 4.** (a) Sensor fabric used in the functional strap textrode sets. (b) Transverse view of the functional strap.

**Figure 5.** Impedance spectra for subject 2. (a) Resistance and (b) reactance.
Cole and body composition parameters

The estimated values for the Cole and body composition parameters are shown in Table 3. This table provides the mean values of the parameters obtained with the Bioimp software from the EBI spectral measurements performed on all three subjects using the textrodes and the Ag/AgCl electrodes.

It is clear that the values of \( R_0 \) and \( R_\infty \) obtained from the textrode measurements are smaller than the values obtained from the Ag/AgCl measurements. The estimation of the characteristic frequencies, \( f_{\text{char}} \), showed small differences, with subject 3 presenting the largest difference, 1.77 Hz.

The body composition parameters also exhibit very small differences. The estimation of the total body water parameter presents a maximum difference of 1.67%, whereas the maximum difference obtained for the ECF and ICF parameter was <1%. The difference obtained for the fat mass and the fat-free mass parameter was 2.29%.

Besides the mean value, the s.d. was estimated for the Cole and body composition parameters. The dispersion of \( R_0 \) and \( R_\infty \) presented in Table 4 shows a slightly higher variability obtained with the textrode in almost all the cases. The characteristic frequency \( f_{\text{char}} \) did not follow any specific trend, but the variability obtained with both electrodes was still low. The body composition parameters present a smaller difference between the two types of electrodes; the data scattering obtained with both sensors are very similar.

**DISCUSSION**

The electrode polarisation impedance \( (Z_{\text{ep}}) \) is produced by displacement of positive and negative electric charges in the skin-electrode interface, and factors such as the absence of electrolytic medium, size of the skin–electrode contact area (electrode contact area) or conductivity of the material in the electrodes could affect the transfers from ion to electron conduction in the skin–electrode contact area. For this reason, the presence or absence of electric charges in the interface will affect the value of \( Z_{\text{ep}} \) and consequently the impedance estimation. The four-electrode configuration is a method that removes the direct influence of the electrode polarisation impedance, but it does not eliminate the influence on, for example, sensitivity to capacitive leakage.

The dry skin–electrode interface commonly present in textrodes seems to result in higher values of \( Z_{\text{ep}} \) and the reason for this could be attributed to the lack of electrolytic medium. It is expected that larger values of \( Z_{\text{ep}} \) contribute to increase the influence of capacitive leakage on the EBI measurement. The custom design of our textrode functional straps appears to reduce the high-frequency artefacts; the dispersion is much lower in the reactance at high frequencies (Figure 6b). The larger sensor area and the more effective and uniform contact produced by the foam layer pushing the textrode surface against the skin may minimise the sensitivity to capacitive leakage and electrode impedance mismatch on the obtained spectral measurements. This can be observed from the impedance spectra in Figure 5, where the spectra obtained with the textrode straps produce minor variations and lower frequency dependency compared with the spectra obtained with previous textile prototypes even in dry conditions.

**Constriction zone and dual opposite electrode measurements**

The injected electrical current flowing in biological tissue is not uniformly distributed; there are volumes with higher current densities and thus a larger contribution to the overall impedance. The area neighbouring the electrode is an area with high current density known as the constriction zone. The larger the area of a current electrode, the smaller the value of the current density through such electrode, decreasing the area of the constriction zone as well. In the case of the textrodes used in this work, the reduced size of the constriction zone produces a more uniform current density distribution. In the same way, the placement and number of Ag/AgCl electrodes have an influence in the effective area for current injection and voltage sensing modifying the constriction zone.

As presented in Marquez et al., modifying the constriction zone and a possible presence of electrode mismatch may have an influence in the spectral impedance measurements. The influence of mismatch may be attributed to differences in the \( Z_{\text{ep}} \) caused by differences in the skin–electrode interface depending on where on the surface of the body the electrodes are placed and/or by the differences of the effective contacting surface of the current and voltage electrodes. As a consequence, the estimated Cole parameters also differ, and this difference is mainly reflected in the values of \( R_0 \) and \( R_\infty \). The measurements done using the dual opposite configuration increase the contact area mimicking better the behavior of the textrode straps. The area increment allows a more representative comparison and at the same time reduces the possibility of impedance electrode mismatch.
Table 3. Mean value of the Cole and body composition parameters

<table>
<thead>
<tr>
<th>Electrode</th>
<th>R₀</th>
<th>R∞</th>
<th>fchir</th>
<th>TBW %</th>
<th>ECF %</th>
<th>ICF %</th>
<th>FM %</th>
<th>FFM %</th>
</tr>
</thead>
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<td>Ag/AgCl</td>
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<td>333.84</td>
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<td>40.21</td>
<td>59.79</td>
<td>23.87</td>
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<tr>
<td></td>
<td>Textrode</td>
<td>518.25</td>
<td>320.40</td>
<td>25.45</td>
<td>57.40</td>
<td>39.86</td>
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<tr>
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<td>Ag/AgCl</td>
<td>525.49</td>
<td>343.36</td>
<td>35.05</td>
<td>54.56</td>
<td>41.93</td>
<td>58.07</td>
<td>25.47</td>
</tr>
<tr>
<td></td>
<td>Textrode</td>
<td>502.73</td>
<td>330.05</td>
<td>35.61</td>
<td>55.94</td>
<td>42.12</td>
<td>57.88</td>
<td>23.57</td>
</tr>
<tr>
<td>Subject 3</td>
<td>Ag/AgCl</td>
<td>531.89</td>
<td>341.52</td>
<td>27.64</td>
<td>61.80</td>
<td>41.26</td>
<td>58.74</td>
<td>15.58</td>
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<td>Textrode</td>
<td>505.40</td>
<td>329.83</td>
<td>29.41</td>
<td>62.99</td>
<td>41.89</td>
<td>58.11</td>
<td>13.95</td>
</tr>
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</table>

Abbreviations: ECF, extracellular fluid; FFM, fat-free mass; FM, fat mass; ICF, intracellular fluid; TBW, total body water. NB The measurements taken with the Red Dot Ag/AgCl electrode are dual opposite electrode measurements.

Table 4. S.d. of the Cole and body composition parameters

<table>
<thead>
<tr>
<th>Electrode</th>
<th>R₀</th>
<th>R∞</th>
<th>fchir</th>
<th>TBW %</th>
<th>ECF %</th>
<th>ICF %</th>
<th>FM %</th>
<th>FFM %</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject 1</td>
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<td>0.70</td>
<td>0.52</td>
<td>0.17</td>
<td>0.13</td>
<td>0.12</td>
<td>0.12</td>
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<tr>
<td></td>
<td>Textrode</td>
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<td>0.67</td>
<td>0.24</td>
<td>0.12</td>
<td>0.12</td>
<td>0.12</td>
<td>0.17</td>
</tr>
<tr>
<td>Subject 2</td>
<td>Ag/AgCl</td>
<td>0.35</td>
<td>0.55</td>
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<td>0.08</td>
<td>0.08</td>
<td>0.12</td>
</tr>
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<td>0.29</td>
<td>0.10</td>
<td>0.11</td>
<td>0.11</td>
<td>0.14</td>
</tr>
<tr>
<td>Subject 3</td>
<td>Ag/AgCl</td>
<td>0.70</td>
<td>0.52</td>
<td>0.17</td>
<td>0.09</td>
<td>0.08</td>
<td>0.08</td>
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</tr>
<tr>
<td></td>
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<td>0.14</td>
<td>0.10</td>
<td>0.08</td>
<td>0.08</td>
<td>0.13</td>
</tr>
</tbody>
</table>

Abbreviations: ECF, extracellular fluid; FFM, fat-free mass; FM, fat mass; ICF, intracellular fluid; TBW, total body water. NB The measurements taken with the Red Dot Ag/AgCl electrode are dual opposite electrode measurements.

Cole and BCA parameters

As previously indicated and shown in Tables 3 and 4, the differences obtained in the impedance spectra and the Cole and the BCA parameters are apparently caused by the differences in the magnitude of the resistance and by a probable slight electrode impedance mismatch present in the Dual opposite measurements. This difference in the impedance spectra, however, produces negligible differences in the estimation of the BCA parameters. As shown in Table 3, the maximum difference obtained in R₀ is 4.9% (in patient 3), resulting in a difference of 770 ml in the ECF. According to Kraemer et al.,24 the minimum threshold to detect relevant clinical changes in fluid status of dialysis patients should be about 1 l. In the same way, the dispersion obtained with the Ag/AgCl electrodes is negligible. Data presented in Table 4 confirm a low electrode impedance mismatch present in the Dual opposite electrode measurements.

CONCLUSION

The results achieved in this study show the feasibility of obtaining good-quality EBI measurements with the functional textrode strap presented.

The impedance spectra showed small differences between the textrodes and the Ag/AgCl electrodes. In addition, the design and manufacturing modifications made to this functional straps result in a clear improvement compared with previous prototypes. The modifications in surface size, the foam layer and the high conductive fibre appear to be the key factors behind the improvement.

The availability of effective textile garments for EBI spectroscopy will be important for the proliferation of EBI-based personalised health-monitoring systems for body fluid distribution, which could have an important role in home-based therapies such as peritoneal dialysis.

Malnourishment is a common condition presented in patients under peritoneal dialysis treatments where bioimpedance has been used to evaluate the nutritional status through BCA.25,26 Under this condition, patients present unusual low serum levels, commonly used as nutritional markers, and other risk factors associated with higher mortality.

In addition, the use of EBI for the evaluation of the nutritional status and body composition in patients with chronic obstructive pulmonary disease,27–29 Alzheimer’s disease30 and other conditions such as hypertension, congestive heart failure, overhydration or water distribution before and after pregnancy31 can also benefit from the integration of textile sensors in health-monitoring systems.

CONFLICT OF INTEREST

FSM is part owner of Z-Health Technologies AB. KL has equity ownership or stock options in Z-Health Technologies AB. JCM declare has no conflict of interest.

ACKNOWLEDGEMENTS

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Textrode-enabled Transthoracic Electrical Bioimpedance Measurements. Towards Wearable Applications of Impedance Cardiography.
Abstract— During the last decades the use of Electrical Bioimpedance (EBI) in the medical field has been subject of extensive research, especially since it is an affordable, harmless and non-invasive technology.

In some specific applications such as body composition assessment where EBI has proven a good degree of effectiveness and reliability, the use of textile electrodes and measurement garments have shown a good performance and reproducible results.

Impedance Cardiography (ICG) is another modality of EBI that can be benefit from the implementation and use of wearable sensors. This technique is based on continuous impedance measurements of a longitudinal segment across the thorax taken at single frequency. The need of a specific electrode placement on the thorax and neck can be easily ensured with the use of a garment with embedded textrodes. The first step towards the implementation of ICG technology into a garment is to find out if ICG measurements can be performed with textile sensors with good enough quality of the signal that could allow the estimation of the fundamental ICG parameters.

In this work, the measurement performance of a 2-belt set with incorporated textrodes for thorax and neck was compared against ICG measurements obtained with Ag/AgCl electrodes. The analysis was based on the quality of the fundamental ICG signals (∆Z, dZ/dt and ECG), systolic time intervals and other ICG parameters. The obtained results indicate the feasibility of using textrodes for ICG measurements with consistent measurements and relatively low data dispersion. This way enabling, the development of measuring garments for ICG measurements.

I. INTRODUCTION

The implementation and use of textile technology and garments for Electrical Bioimpedance (EBI) measurements have been studied before, with reported results that are consistent and encouraging. Nonetheless, applied research regarding measurement performance of the garments is still necessary.

Previous studies of bio potential sensing [1, 2]and EBI for Body Composition Analysis have shown the feasibility of using, textile electrodes, textrodes[3, 4]. In Impedance Cardiography the use of a measurements garment with integrated textrodes would facilitate continuous monitoring of cardiac activity.

The usefulness of ICG has been questioned by many but it is endorsed by the U.S. Department of Health & Human Services through the Centers for Medicare & Medicaid Services (CMS) [5] for a number of clinical uses, patient management and monitoring applications. And the particular and distinctive capability of ICG to provide beat to beat cardio-vascular information non-invasively opens for several monitoring scenarios.

It is particularly some of the applications endorsed by the CMS, that would benefit greatly from a wearable-ICG measurement system. One example is optimization of fluid management in patients with congestive heart failure based on patient monitoring at home.

Fig.1 A) Measurement set-up. B) Textile-belts for neck and chest with the four textrodes highlighted in red.

This work was supported in part by the Mexican Conacyt under Scholarship 304684.

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In a preliminary study, ICG and ECG signals were recorded with the textile belts showing waveforms similarities. For this study the analysis of the textile belts was extended to a performance comparison of some parameters calculated from the characteristic points of the recorded signals i.e. heart rates, peak first time derivative, time intervals, and stroke volume (SV). The comparison was done using the thoracic electrical impedance (TEB) measurements obtained with Ag/AgCl electrodes as reference.

II. IMPEDANCE CARDIOGRAPHY

The biophysical property of blood, double of the conductivity compared to muscle tissue and several times higher than other type of tissue [6, 7], produces cardio-related impedance changes that are easily recorded. Important parameters reflecting cardiac activity can be obtained from thoracic electrical bioimpedance (TEB) measurements such as impedance change $\Delta Z$ and a first time derivative $dZ/dt$. Using the correlation found by Lababidi et al [8] between the ICG waveforms and mechanical events of the cardiac muscle, characteristic points from the ECG an ICG signals can be used to estimate systolic time intervals and other clinically important hemodynamic indices.

ICG technology however, has not reached a full consensus among practitioners in the diagnosis and monitoring of hemodynamic activity in patients with cardiovascular conditions. Under clinical criteria, this technology has not been awarded as an established method, primarily due to: it has presented an unpredictable accuracy in the hemodynamic estimation for patients with severe heart disease or critically ill [9] and secondly the controversy behind the physiological and anatomic origin of $dZ/dt$ [10].

Nonetheless the United States Department of Health and Human Services (USHHs) endorses the use of TEB for specific cases of the following medical conditions [5]:

- Optimization of atrioventricular (A/V) interval for patients with A/V sequential cardiac pacemakers.
- Monitoring of continuous inotropic therapy for patients with terminal congestive heart failure.
- Evaluation for rejection in patients with heart transplant as a predetermined alternative to myocardial biopsy.

III. MATERIAL & METHODS

A. Textrode Belt

A set of two custom-made textile belts for neck and chest have been manufactured with neoprene and synthetic wrap knitted fabric at Swedish School of Textiles at the University of Borås. Four textrodes were embedded in the belts following a lateral spot electrode array [11] as is shown in Fig. 1.B. The textrodes had a contact area each of approximately 13.5 cm$^2$ and are made with conductive Velcro with loops made of silver coated fibers. Snap-buttons were used to enable the connection with the ICG measurement instrumentation. The right adjustment of the belts and the correct positioning of the textrodes were achieved through Velcro fasteners and the stretchy properties of the belts materials.

B. Ag/AgCl Electrolytic Gel Electrode

RedDot repositionable electrodes manufactured by 3M were used in this study. The rectangular electrodes with a contact area of 10.1 cm$^2$ and snap-button connectors, utilize conductive and adhesive hydro gel to ensure the right contact.

C. ECG and ICG Measurements

For this study a TEB-measurement of one-minute duration was recorded on three healthy subjects standing in a resting state and following the measurement protocol presented in Fig.2. Using the texttrode belt set and
conventional Ag/AgCl electrodes, single-point frequency measurements at 50 kHz were performed with the impedance cardiograph Respimon. The measurements were done using a tetrapolar configuration and the lateral spot electrode array as suggested by Woltjer in [11]. The measurement set-up used is represented in Fig. 1A).

D. Measurement Instrumentation

**Impedance Cardiograph Respimon.** - Contains an impedance plethysmography device that provides 2 EBI analogue outputs: base impedance $Z_0$ and impedance change $\Delta Z$ from a tetrapolar TEB impedance measurement performed at 50 kHz and a biopotential amplifier that produces an ECG signal.

**LabView Data Acquisition Card NI USB-6218.** - This card was used to digitalize the analogue signals produced by the impedance cardiographer. The Card was connected through USB to a PC/compatible that records and storage the measurements for further processing and analysis with Matlab.

E. Measurement Analysis and Studied Parameters

**Matlab Processing.** - To remove unwanted noise such as offset, 50 Hz noise and respiration a digital filtering stage was implemented. Time delay estimation and compensation for the digital filters, signal synchronization, characteristic point detection, parameter estimation and statistical process was also done in Matlab.

**Analyzed Parameters.** - Heart Rate estimated from the ICG and ECG and peak first time derivative $dZ/dt_{max}$ were some of the parameters analyzed in this study. In addition, relevant hemodynamic stroke volume (SV) index and timing parameters Left Ventricular Ejection Time (LVET) and R-Z were also compared.

**Left Ventricular Ejection Time (LVET).** - Is the time elapsed from the opening to the closure of the aortic valve. In figure 3 correspond to the time between spots B and X.

**R to Z time (R-Z).** - Represents the time span between R-peak in the ECG and the maximal point in the $dZ/dt$ curve. In figure 3 correspond to the time between R and Z spots. R-Z time is used in the estimation of the Heath Index[12, 13], considered an index of the cardiac contractility.

**Stroke Volume Formula.** - The estimation of Stroke Volume was done using the equation 1. In this equation proposed by Bernstein *et al* [14, 15], $V_{EPT}$ represents the volume of electrically participating thoracic tissue (mL), $dZ(t)/dt_{max}$ the peak first time derivative (ohm s$^{-2}$), $Z_0$ the measured transthoracic base impedance (ohms) and LVET the left ventricular ejection time (s).

$$SV = V_{EPT} \frac{dZ(t)/dt_{max}}{Z_0} LVET \quad \text{Equation (1)}$$

**Performed Comparison.** - The comparison in this work was done based on the signal quality of the ICG and ECG curves and on consistencies in the estimation of relevant parameters. The signals were recorded during one minute and the parameter variation was analyzed by means of mean values and data dispersions.

IV. MEASUREMENT RESULTS

A. ICG Recordings

In Fig. 4 the fundamental ICG signals impedance change $\Delta Z$, first derivative $dZ/dt$ and ECG corresponding to subject 1 are presented. From these signals no significant dissimilarities can be observed. The R-peak magnitudes acquired with the textile belts seem to be slightly larger, although this could be attributed to variations of the electrode placement with the two different types of electrodes.
B. ICG Parameter Estimation

The estimated heart rate values obtained from the impedance and the ECG signals are presented in Fig. 5. In this figure, the HRs obtained from the electrolytic electrode were analyzed and compared against the HRs obtained from the textile belts. From figure 5A) and 5B) high degree of similarity and minor differences were observed between the HRs derived from the ECG and ICG signals. Both types of electrodes, textile and electrolytic, did not show significant differences in mean values and dispersions for any of the subjects.

The results of the time interval LVET and the R to Z time are presented on Fig. 6. For the LVET in Fig. 6A), a clear overlap can be observed for the values obtained from the three subjects with a maximal mean difference of 39.31 ms for subject two. The largest dispersion is observed with the Red Dot electrode for subject 2. In the R to Z period shown in Fig. 6B, noticeable overlapping between the two types of electrodes was obtained for the three subjects. Subject 2 presented the most evident mean difference of less than 10 ms whereas the largest data variance was observed in the subject 3 when using the textile belt.

The peak first time derivative $\frac{dZ(t)}{dt}$ and the SV estimation are shown in Fig. 7. The SV estimation depicted in Fig. 7A) presented overlapping zones for the three subjects and a very close mean value for subject 3. Subject 2 exhibited the largest mean difference of 13.37 ml and the largest data dispersion in the RedDot Ag/AgCl electrode. The peak first time derivative $\frac{dZ(t)}{dt}$ also present evident overlapped zones for all the subjects. In Fig. 7B) is possible to observe how the largest mean difference is found on subject three

<table>
<thead>
<tr>
<th>Source</th>
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<td>ICG</td>
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V. Analysis & Discussion

The study did not aim to quantify the validity or precision of the parameter for any specific medical condition but to evaluate the performance of textrodes for ICG based on the estimation of relevant parameters and ICG curves.

A. ICG Curves and HR estimation

The ICG waveforms shown in Fig. 4, in general, exhibit good resemblance. Differences in the amplitude of the R-peaks depend more on the exact position of the electrodes than the electrode type. The time shifting observed are caused by the natural heart rate variability rather than by the electrode difference. The HR analysis presented in Table I was done with respect to the HR obtained from the ECG.
From this table is possible to observe a HR mean value obtained from both sources, ECG and ICG. The small differences observed are most probably attributed to the natural HR variability of different measurement sessions and not necessarily to the type of electrode. There were no missing beats and no need of special signal processing for the signals acquired with belts, thus the same algorithms for processing and detection were used for both types of electrodes.

B. Stroke Volume equation

Stroke Volume (SV) is probably one of the most medical relevant hemodynamic indices obtained using Thoracic Electrical Bioimpedance TEB [16, 17]. Despite the clinically attractive attributes of SV estimation by means of TEB, the method has shown low consistency between the impedance wave forms and hemodynamic variables [18]. This inefficiency in the correlation between curves and estimated variables suggest a lack of robustness in the equation models to estimate SV and a low sensitivity and specificity of the impedance signal acquisition in some medical conditions. In addition to this, a need of consensus whether the physiological origin of dZ/dt comes predominantly from volume changes as traditionally believed or whether the main contribution comes from blood resistivity changes in the vessels as more recent evidence was presented by Bernstein et al. [19, 20] might also influence. In this proposed model called Electrical Velocimetry the excess of extravascular lung water it is also taken in account.

C. Stroke Volume & dZ(t)/dt_{max} estimation

The peak value dZ(t)/dt_{max} is an important parameter in the estimation of stroke volume and other indices. The peak first derivative measurements done with the belts presented a reasonable level of consistency. As it can be seen from Table II, measurements performed with the textrode belts, showed a mean variation from 88.4% to 114.8% of the values obtained from the Ag/AgCl reference electrode. Similar behavior was observed for the SV estimation where mean values of the textrode belts showed differences from 96.68% to 120.4% of the values acquired with the reference. No evident trends related with the type of electrode were found in the estimation of these two parameters. However, since measurements with the two types of electrodes were not taken simultaneously rather sequentially in a different measurement session, variations in the SV estimation may not be entirely attributed to the type of electrode used. Variations may also be due to the natural biological differences between two measurement sessions.

D. Ejection & R to Z time

The timing parameters estimated with the textrode belts presented relative close resemblance. The LVET values presented in Table II showed reasonable values that range from 341 to 360 ms. Similar measurements were obtained and reported by Cokkinos et al. [21]. Indeed the values obtained with the belt represent 92.2% to 112.6% of the value measured with the corresponding Ag/AgCl electrode. Alike performance was observed in the estimation of R to Z time where measurements fluctuated from 97.2% to 105.9% of the corresponding value. This difference seems to be very small and although such timing parameters can be used to estimate hemodynamic information, the influence of observed differences on the estimation of such hemodynamic indices, was not analyzed in this study.

VI. CONCLUSION

As a first step to evaluate the performance of textile electrodes for TEB measurements to obtain cardioimpedance recordings, the analysis of ICG recordings, time period LVET, characteristic period R to Z and left ventricular stroke volume performed in this work, establishes a good starting point towards a more robust assessment of the use of textile sensors.

Determining if the parameters estimated with the textrode belts produce accurate and reliable enough results is highly dependent on the clinical application or individual case, for instance an error rate of 10% in the HR may be well acceptable for some applications while in others one, 0.5% may be not acceptable for an accurate diagnostic or may not be useful to differentiate between healthy and disease. In any case for both the waveforms and the calculated parameters the resemblance is very close.

In spite of the consistent results and apparent good and reliable estimation of parameters confirmed by the analysis of the timing and systolic parameters done in this study, further analysis is required in order to determine the implications of using textile electrodes for the determination of hemodynamics indices.

**REFERENCES**


