

Chapter 2

Wearable Bioelectric Signal Acquisition

Bioelectric signals or biopotentials are generated by nerves and muscles and embody the activity of particular organs: the heart, brain and muscle [1, 2]. The continuous acquisition of these physiological signals allows to detect and prevent the progress of certain diseases such as cardiovascular diseases or neurological pathologies. In addition, it also has the potential to support the rehabilitating and chronic ill patients. Bioelectric signals are obtained through specific electrodes that establish an interface between the human body and the measurement apparatus [3]. In order to design readout circuits to measure bioelectric signals and to provide solutions for real-time monitoring, it's necessary to cope with various problems due to particular characteristics of these signals, as well as with environmental and device-related interferences. Therefore, the design of wearable bioelectric acquisition systems

requires a solid understanding of the origin and characteristics of bioelectric signals as well as the system components and design.

This chapter will focus on introducing the origin and principle of bioelectric activity as well as the measurements and acquisition system involved particularly in detecting the electrocardiogram (ECG), the electroencephalogram (EEG), the electromyogram (EMG), and the electrooculogram (EOG).

2.1 Bioelectric Signals

In order to fully understand the nature and characteristics of bioelectric signals it's necessary to explain the basics of bioelectricity phenomena and how these signals are originated. There are different types of bioelectric signals, depending on the organ or function they are associated with. All these points are explained in detail further along this chapter.

2.1.1 Origin

Bioelectricity is a phenomenon existent in many living element (cells, tissues, organs) and provide both steady and time-varying electric potentials that represent certain functions of organs such as heart, brain and muscles. Biological tissues can be considered as electric volume conductors, supporting the conduction of currents [4]. On a larger scale, few places in the body are non-conductors, which reflect the little amplitude variance occurred from one part of the body to another. Therefore any current generator within the body can create electric fields that can be acquired from most parts of the human skin, usually called bioelectric signals [1].

Bioelectric processes occur at the cellular level resulting in segregation of charge and thereby electric fields within the body. These cells are called excitable cells and when stimulated they undergo depolarization, giving origin to action potentials . The occurrence of this phenomenon is accompanied by physiological events such as transmission of information along nerve cells or the contraction of cardiac cells [1, 5]. Figure 2.1 shows the action potential generation mechanism along with the structure of a cell membrane.

A single excitable cell exhibits a resting potential of around 70mV with respect to the extracellular medium [1, 4]. At this state, the membrane of the cell is more permeable to K^+ than Na^+ , with higher intracellular concentrations of K^+ . The transport of these ions is made through cell molecular pumps and selective ion channels (Figure 2.1).

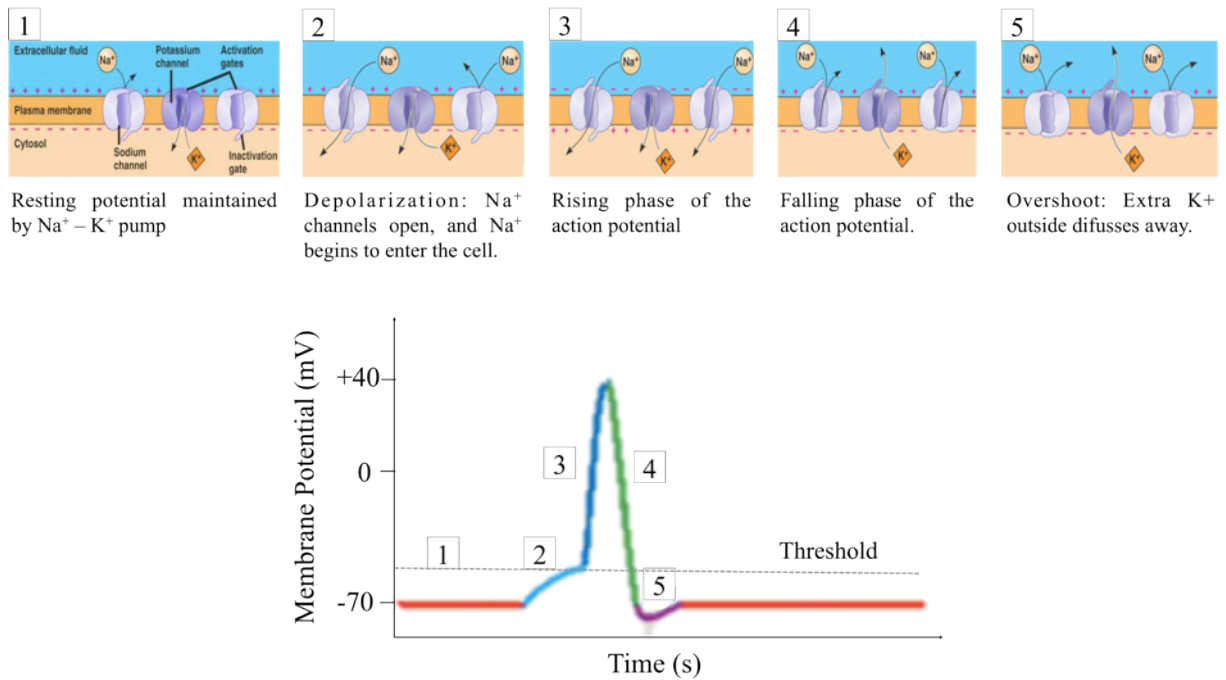


Figure 2.1 Action potential generation mechanism. Each step is represented in the action potential plot, as a colored region.

When a cell is electrically stimulated and exceeds a certain threshold value (typically of 20 mV), the membrane potential starts a rapid depolarization due to a change in permeability towards the increase in Na^+ ions. This causes the Na^+ ions to diffuse inwards the cell and results in a potential increase in the interior of the cell. When the potential reaches a value close to 40 mV, the Na^+ ion permeability starts to increase more slowly, allowing the ions to flow from inside to outside, returning the membrane potential to its resting value [1, 2, 4].

An action potential corresponds to this cycle of cellular potential (Figure 2.1), and resultant generated currents propagate themselves giving origin to bioelectric signals, such as ECG, EEG, EMG and EOG [1, 5].

2.1.2 Main Bioelectric Signals

Each excitable cell produces a characteristic action potential, that depending on propagation and location, giving rise to different bioelectric signals. For example, the activity of cells of a massive number of neurons results in EEG signal, activity of cells in the sinoatrial node of the heart produces an excitation that when propagated throughout the heart results in ECG. Thus, it is clear that depending on the type of cell, different bioelectric signals are produced, with distinct characteristics and measurement procedures [4].

Despite the existence of more bioelectric signals, ECG, EEG, EMG and EOG are the most important considering a wearable monitoring context. In addition, the detection of these bioelectric signals is performed non-invasively, i.e. on the surface of the skin.

ECG

The first findings of heart bioelectrical phenomena occurred back in 1842, when Calo Matteucci (Italian physicist) found that each heartbeat is accompanied by an electric current [6]. Since then, a lot of effort has been put into ECG research. An ECG is a recording of bioelectric signals originated from cardiac electric activity, usually measured by placing electrodes directly on the body [7]. This activity is known to reflect the activity of the heart muscle underneath and in its proximities.

The heart comprises four types of tissues: sinoatrial node (SA) and atrioventricular node (AV), atrial, Purkinje, and ventricular tissue. These tissues are composed of excitable cells exhibiting its own characteristic action potential [1, 7]. Figure 2.2 depicts the heart anatomy and the bioelectric events occurred during an ECG.

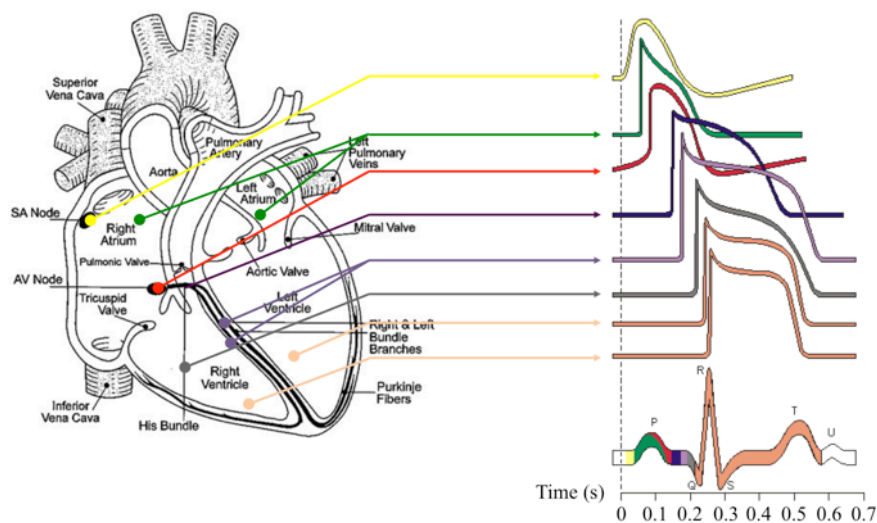


Figure 2.2 Heart anatomy and major bioelectric events of a typical ECG.

Cardiac electric activity starts at the SA node and is then conducted to the ventricles. The complete ECG is shown in Figure 2.2 and it can be divided in three components, each one corresponding to a specific electrical activity phenomena: P wave, QRS complex and T and U waves. The P wave corresponds to activation or depolarization of the atrial cells, arising from the SA node. Following this wave, an isoelectric segment (P-R segment) appears preceding a rapid and large deflection that corresponds to the excitation of ventricles – QRS complex. This complex begins with a descending deflection, the Q wave, headed by R wave (upward deflection) and ending with a downward deflection, the S wave. Finally, ventricles

return to their electrical resting state showing a low-frequency T and U waves that indicate the ventricular repolarization. This series of bioelectric events form a cardiac cycle, i.e. heartbeat, and being the normal heart rate comprised in the range of 60 to 100 beats per minute. Common abnormalities detected in ECG are identified through the analysis of these waveform components and examples are: absence of P waves, fast or slow heart rates and non isoelectric ST segments [1, 7, 8].

Physically, the simplest model for linking the cardiac generator to the body surface potentials and provide a framework for the study of clinical ECG is the dipole model. Therefore, heart's activation is an electric vector usually called cardiac equivalent vector (C), which can be measured if using a differential recording [9]. Basically, two electrodes are placed on the body forming a lead (L_{AB}) between them, and the potential difference between them, measured on the surface is:

$$V_{AB}(t) = C(t) \cdot L_{AB}(t), \quad (2.1)$$

where A and B represent both measurement locations. This concept of leads was first introduced by Willem Einthoven in 1902, when he proposed a measurement convention named after him – Einthoven lead system [10]. The approach comprises a combination of electrodes taking measurements from different leads: limb and chest leads. Figure 2.3 translates the Einthoven's assumption that the heart is the electric center of a triangle defined by the leads – the Einthoven triangle [7].

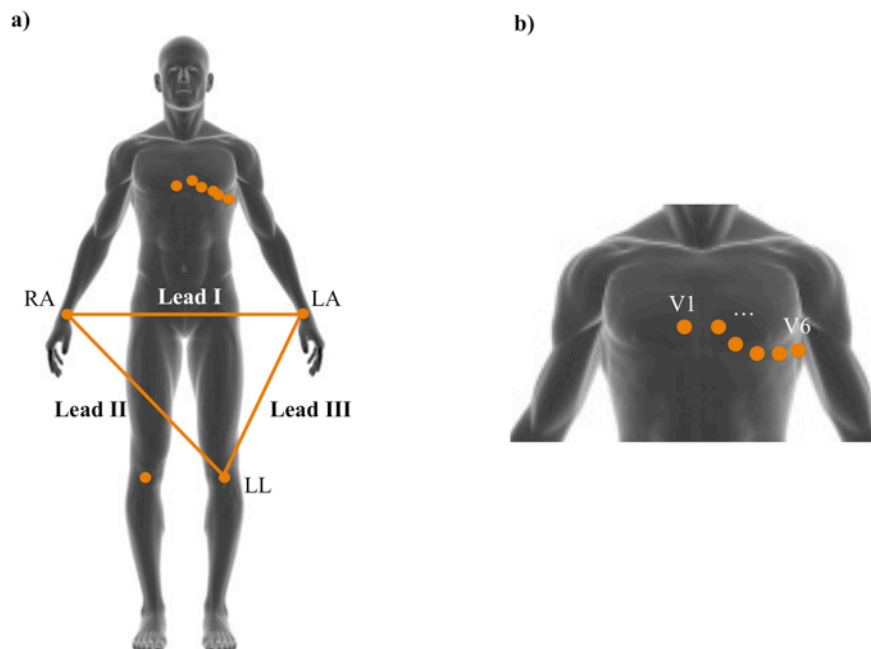


Figure 2.3 Einthoven lead system: a) limb leads, and b) chest leads (leads are incrementally numerated from V1 to V6).

According to this methodology, an ECG is obtained through the derivation of three limb electrodes, i.e. leads, and their potentials are called lead I, II and III. Each one of these leads is defined as:

$$I = V_{LA} - V_{RA} \quad (2.2)$$

$$II = V_{LL} - V_{RA} \quad (2.3)$$

$$III = V_{LL} - V_{LA} \quad (2.4)$$

where subscript RA = right arm, LA = left arm, and LL = left leg. Einthoven's leads fulfill an electrical outlook of the heart from three different vectorial directions. In addition to these limbs, unipolar leads aVR, aVL and aVF can be used to record the potential at the electrode placed in the right arm, left arm and left foot, respectively. The remaining six leads V1, V2, V3, V4, V5 and V6 are designated as chest leads and together with the other leads contribute to define the nature and status of the activity on a specific part of the heart muscle. For instance, inferior myocardial infarction produces main changes in the leads that explore the heart from below, i.e. leads II, III and aVF. At ECG frequencies (0.05 – 150 Hz), the human body is assumed as merely resistive, allowing to consider the four limbs as wires attached to the torso. Therefore it's possible to record a lead in different locations of the limb, without loss of cardiac information. Nevertheless, there is a signal magnitude variation that is induced by different inter-electrode distances and locations [7].

In a study performed by Merja Puurtine and co-workers it was shown that ECG amplitude is affected by the inter-space electrode distance. For instance, the recorded amplitude for the electrode pair V2–V6 and V1–V2 were, respectively, 3.711 mV and 1.401 mV [11]. Therefore, higher amplitudes are obtained with longer inter-electrode spacing. Despite this, there is a point where the distance from the heart, influences negatively the amplitude of the ECG signal. According to [12], large voltages are recorded in the precordia leads in comparison with the unipolar limb leads.

Clinical interpretation of ECG is useful in many applications including diagnosis of arrhythmias, ischemia, myocardial infraction, and so on. However, proper instrumentation and technical specifications are required and have been proposed by the American Heart Association and the Association for the Advancement of Medical Instrumentation.

EEG

The Austrian psychiatrist Hans Berger was the first one to record the human EEG in 1929, and since then this bioelectric signal has been the most utilized to clinically monitor brain function [13]. An EEG is a superposition of many different bioelectric sources in the outer cortex that generate measurable oscillations of brain electric potential from the human scalp [14, 15]. These signals are generally difficult to decode since they translate the activity of billion of neurons diffused via brain tissues, fluids and scalp. However, EEG is still a useful tool to detect pathologies such as brain tumors, epilepsies, infectious diseases, head injuries and sleep and metabolic disorders [16].

The brain is a complex organ with massive bioelectrically active neurons and with three primary divisions: brainstem, cerebellum and cerebrum. The largest part of the brain is the cerebrum, and can be divided into the right and left hemispheres, each relating to the opposite side of the body. The surface layer of each hemisphere is called the cerebral cortex, containing about 10^{10} nerve cells (neurons) and believed to generate most of the electrical activity measured on the scalp. The cortex represents the processing unit for sensorial and motor signals, receiving sensory information from the skin, eyes, ears and other receptors [15, 17]. There are four functional sub-divisions or lobes of the cerebral cortex, as shown in Figure 2.4.

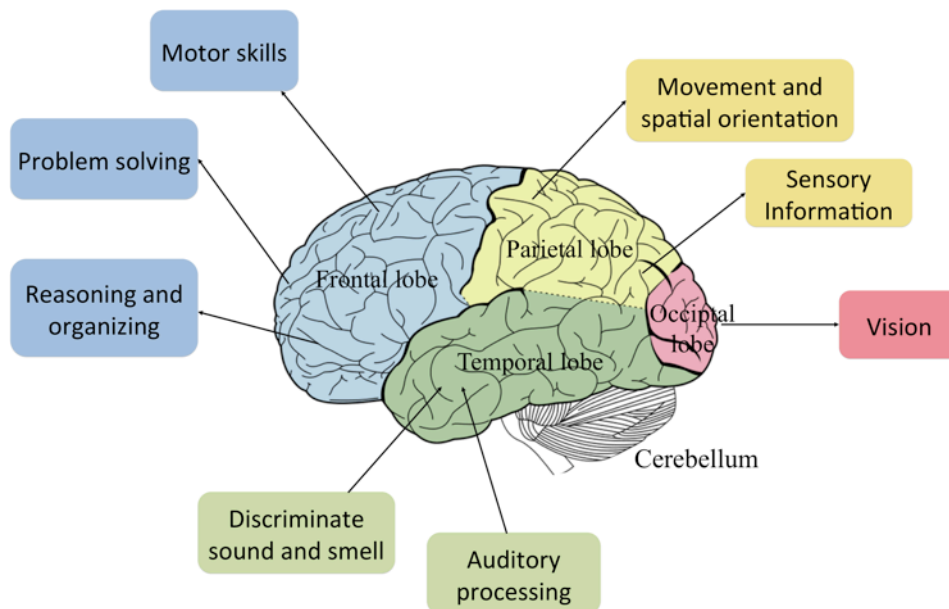


Figure 2.4 Brain main lobes and associated functions.

As shown in Figure 2.4, the fissures are the major dividing landmarks of the cerebral cortex resulting in four lobes: frontal, occipital, parietal and temporal. Each one of these lobes can be connected with a different function such as auditory, motor or visual. The front part of

the brain is called frontal lobe and is involved in reasoning, motor skills, organizing, problem solving and a variety of higher cognitive functions such as behavior and emotions. The visual system is mainly controlled by the occipital lobe and is located at the back portion of the brain. This lobe is responsible to interpret visual stimuli and information from the eyes. The parietal lobe is associated with the integration of sensory information from different parts of the body and is located in the middle section of the brain. General functions of this lobe include movement, spatial orientation, speech, pain and touch sensation. The bottom region of the cortex is called the temporal lobe, and can be divided into two parts, each located on both sides of the skull. The temporal lobe is responsible to coordinate auditory processing, interpreting sounds and language, as well as to distinguish and discriminate smell and sound [17, 18].

Electrical activity measured in the scalp can be divided into two types: spontaneous potentials (example: beta or alpha rhythms) and evoked potentials or event-related potentials [15]. The latter is the direct response to some external stimulus like an auditory tone or a visual signal, whereas event related potentials are dependent on the brain processing of the stimulus. Properties such as frequency, amplitude and recording site are often used to characterize spontaneous EEG waveforms. EEG spectral analysis allows to associate each pattern with certain mental states such as sleep or consciousness [16]. Major brain rhythms are categorized according to their predominant frequency components and can be classified as: alpha, beta, delta, gamma and theta waves. Figure 2.5 shows frequency characteristics and mental states associated with each EEG waves.

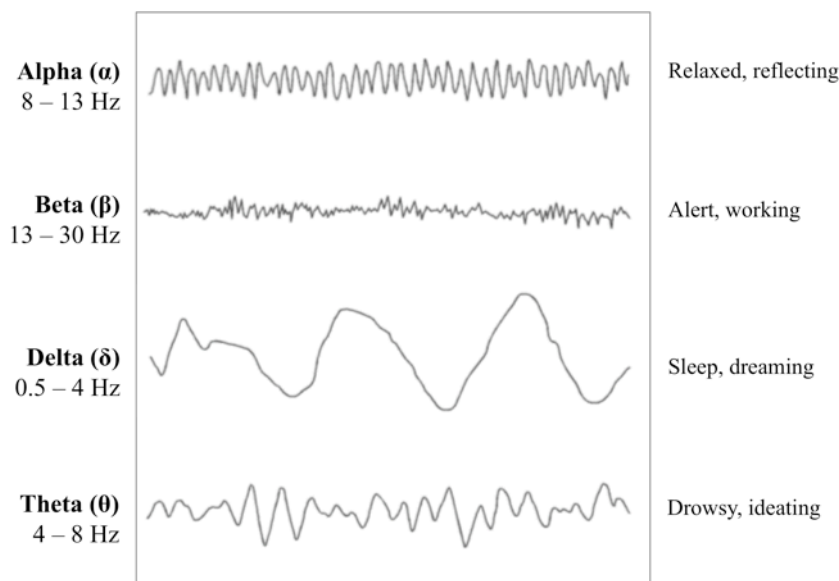


Figure 2.5 EEG brain waves according to different states of consciousness (adapted from [16]).

There is a progression of EEG activity from a state of wakefulness to deep sleep, which reflects mainly on a decrease of frequency and increase in amplitude. Alpha rhythms are characterized by frequencies of 8 to 13 Hz and typical of an awake, quiet and resting state of consciousness. These waves have higher amplitudes on occipital and frontal areas of the brain, being the typical value below 50 μV in adults. On the other hand, beta waves have smaller amplitudes (20 μV) but higher frequency components ranging from 14 to 30 Hz. These waves are more frequently recorded from the parietal and frontal regions of the brain and are particularly present during intense mental activity. There's also a higher-frequency EEG wave called gamma that is characterized by frequencies above 35 Hz and amplitudes of 3 to 5 μV . Gamma waves are usually accompanied by sudden sensory stimuli. Waves from 4 to 7 Hz are called theta waves and occur mainly in the parietal and temporal lobes. These waves have amplitudes of 20 to 100 μV and are typical of complex behaviors such as learning and memory. As for delta waves, these have standard amplitudes of 20 to 200 μV and frequencies below 3.5 Hz. Delta waves occur in deep sleep, coma or serious organic brain diseases [15, 16, 19].

Similarly to ECG, EEGs are also recorded according to a lead system that includes several electrode's location around the subject scalp called 10-20 lead system (Figure 2.6).

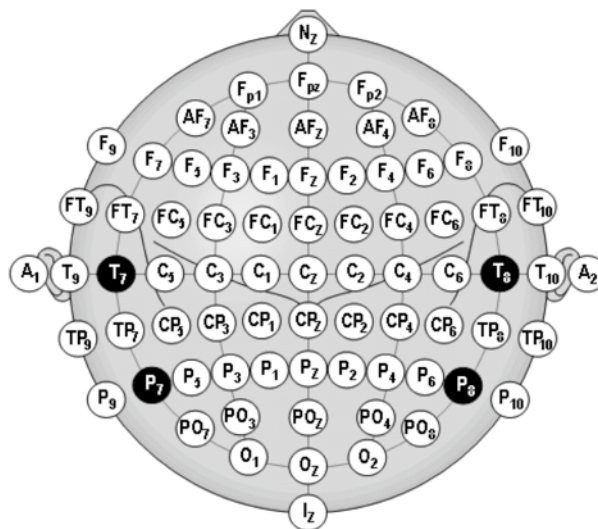


Figure 2.6 International 10-20 system of EEG electrode placement [16].

Electrodes are labeled by letters according to their positions on the scalp, i.e. depending on the monitored brain region (e.g. frontal or occipital). The 10-20 lead system consists on a diagnostic and preventing tool widely used in the study of sleep patterns, effects of various pharmaceuticals on sleep, epilepsy among others. The electrode selection influences the magnitude of the signals recorded, which ultimately influences the required sensitivity of the

sensor. In a study developed by Charles Epstein and Gail Brickley, it was found that EEG amplitude increased monotonically until a maximum inter-electrode distance of 15 cm [20].

EMG

Bioelectric activity of muscles or myoelectric activity (EMG) was first measured in 1890 by Marey [21]. EMG is generated by activation of muscles prior to contraction and is a result of the summed action potential of individual muscle motor units (MU). Since each muscle contraction involves a large number of cells, the bioelectric current flowing through the fibers gives origin to skin potentials in the range of millivolts [22].

Skeletal muscles are composed by thousands of muscle fibers that are defined as a complex multinucleated cell of variable length (from mm to cm). Muscle fibers are arranged in a parallel configuration to one another and bundle together by connective tissue, which is responsible for providing support and unity of action. MUs comprise the functional units of a muscle contraction and are composed by a group of muscle fibers innervated by one motor neuron [21]. When a neural signal is sent to a motor unit, each MU is contracted resulting in a synchronous activation of all the innervated muscle fibers. EMG signals represent the spatio-temporal summation of this electrical activation of the mechanical system of muscle fibers. These signals represent the level of activity of a specific muscle and are characterized by a stochastic noise assuming a Gaussian distribution function [1]. Figure 2.7 shows that EMG can be related with the strength of an intentional muscle contraction and respective force.

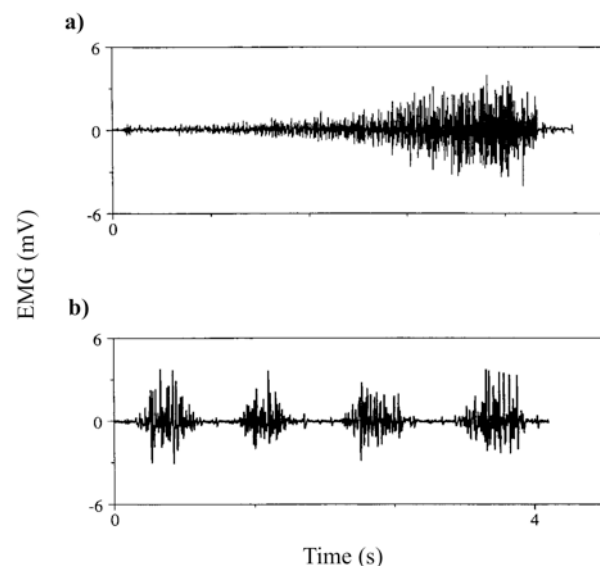


Figure 2.7 EMG signals from a) a static contraction and b) a series of contraction and relaxation [21].

EMG signals are recorded using surface electrodes placed near the muscle groups, preferably between a motor point and the tendon insertion, or between two motor points.

Electrodes should be aligned in a longitudinal midline of the muscle, being this axis parallel to the fiber length. An instrumentation or operational amplifier can be used to perform differential acquisition, similarly to ECG. EMG signals can be related with the applied muscle force. For instance, at muscle fatigues the frequency spectrum of EMG signals shifts towards lower frequencies and has smaller amplitudes. However, its frequency and amplitudes manifest minor changes over a range of low contractile force and progressive large force. According to several studies, an increase in the inter-electrode spacing produces an increase in the EMG medium magnitude [23, 24]. Although this results in difficulties in signal analysis, EMGs are still widely used as a monitoring and diagnostic tool of neuromuscular diseases (eg. Myopathy). In particular, EMG frequency-spectrum analysis finds applications in biomechanics research in order to design controlled prosthetic devices or to detect the degree of muscle fatigue and performance.

EOG

The movement of the eyeballs within the conductive environment of the skull gives origin to an electrical potential – EOG. In order to understand the generation of this bioelectric signal, the eyeballs are considered as dipoles, and electrodes are placed on each side of the eyes, above or below them. Therefore, EOG represents the dipolar current flow from the cornea to the retina, which allows to estimate the eye's angular displacement. Figure 2.8 shows an example of an EOG taken from a healthy subject [1].

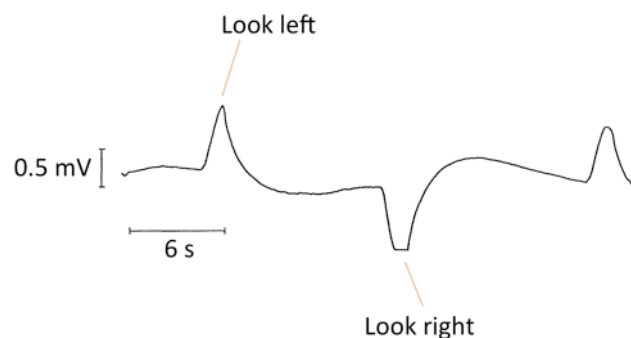


Figure 2.8 Example of an EOG signal obtained with three electrodes [1].

Figure 2.8 shows the clear positive and negative signal peaks that represent the blinking of the eyelids. Clinical applications of EOG include study of disorders of eye movement and balance, sleep and dream research, visual fatigue and evaluation of reading ability. In addition, EOG could also be used in wearable devices for instance in activity recognition and context-awareness [4].

2.2.3 Bioelectric Signals Main Properties and Challenges

Measurement of bioelectric signals involves recording very low voltage and low frequency signals, with high impedance sources, overlaid with interference and noise signals. Essentially, bioelectric signals are associated with various forms of energy and can be characterized as a function of time and space [1, 4]. Therefore, this allows for a non-invasive acquisition of such signals providing vital clues as to normal or pathological functions of organs. Table 2.1 lists the most important bioelectric signals measured from the body, as well as the significant properties.

Table 2.1 Types of bioelectric signals and main characteristics [1, 2, 7, 14, 19].

Bioelectric signal	Biological Source	Amplitude	Frequency
Electrocardiogram (ECG)	Heart	0.5 – 4 mV	0.05 – 150 Hz
Electroencephalogram (EEG)	Brain	5 – 300 μ V	0.5 – 150 Hz
Electromyogram (EMG)	Muscles	1 – 10 mV	0 – 10 kHz
Electrooculogram (EOG)	Eye dipole field	10 – 100 μ V	0 – 10 Hz
Electroretinogram (ERG)	Eye retina	0 – 900 μ V	0 – 50 Hz
Electrocortigram (ECoG)	Exposed surface Brain	-	100 Hz – 5kHz
Electroneurogram (ENG)	Nerve blunder	5 μ V – 10 mV	100 Hz – 1kHz
Evoked potentials	Brain	0.1 – 20 μ V	-
Action potentials	Nerves and muscles	-80 – 80 mV	10 – 10 kHz

Signal Amplitude and Power

Most demanding signals, such as ECG, EEG and EMG are within the μ V range, often going from 5 μ V to 10 mV. Giving such small amplitudes, it is very easy to have a few millivolts superimposed on the measured bioelectric signal, mainly due to power-lines. This is a major problem since magnitude and power of both signals is in the same order (Table 2.1). Likewise, other bioelectric signals lie in the same range of amplitude, resulting in further interference among signals. As an example, ECG or even EOG signals usually appear overlapped on EEG signals. Bioelectric signal amplitudes presented in Table 2.1 represent the values obtained for surface detection, and near the place or source that they are originated. In general, the human body may be considered as a volume conductor, which makes possible to

detect some bioelectric signals in different places in the body. For instance, an ECG can be detected by placing the sensors near the subject wrists, despite the compromise of reduced signal strength when compared with signals obtained near the heart. In addition, since bioelectric signals are measured as a difference of potential between two points, the distance between them interferes with the magnitude of signals detected. Therefore, the design of wearable bioelectric systems requires proper location selection for the measurement electrodes. For instance, the inter-electrode distance influences the bioelectric signal strength, i.e., amplitude.

The maximum power transfer occurs when the source impedance equals the input impedance of the measurement device. In this case, impedance matching occurs. For complex impedances, matching occurs when the conjugate are equal in magnitude. However, since bioelectric signals are within the μV range, it's important to maximize also the voltage produced in the high-impedance load. The main problem with bioelectric signal acquisition is their low power due to the small source currents. This is a problem mainly when implementing power line noise cancelation. The interference is canceled, but the bioelectric signals are also attenuated. This means that any small current flowing to the measurement apparatus will lead to a voltage drop on the transducers, reducing further the available output voltage.

Signal Frequency

From Table 2.1, it's perceptible that bioelectric signals are not difficult to measure regarding spectral components. In fact, maximum frequency is on the order of a few kilohertz. The main problem is related with smaller frequency components, close to DC, which is severely influenced by $1/f$ noise (or pink noise). This noise is inversely proportional to frequency. In addition, bioelectric signals have overlaying spectral components, specially centered in the range of 1 to 100 Hz, causing mutual interference between them. Even simple patient movement, which occurs on the order of a few Hz, interferes with signals such as ECG and EEG. Another common problem is associated with electromagnetic fields coming from power-lines (50 – 60 Hz) that are easily coupled through the power source or by the human body working as an antenna. This coupled signal usually has higher amplitude than the bioelectric signal being measured, which leads to the need to remove the effect of picked-up interference.

2.2 Standard Bioelectric Signal Acquisition System

The phenomenon of bioelectricity involves ions as charge carriers and its recording deals with the transduction of these ionic currents into electric currents. This type of interface is carried out by surface electrodes, consisting of electrical conductors in contact with the aqueous ionic solutions. Different electrodes are used for the recording of bioelectric signals, based on specific transduction schemes: wet, dry and capacitive electrodes. This section is going to focus on the principles of bioelectric transduction and electrode design.

Surface bioelectric signals are small in amplitude due to the impedance barrier created by the electrode-skin interface, leading to more susceptibility to artifacts. These artifacts are a result of the relative motion of the electrode and the skin, the activity of the nearby muscles and other instrumentation and environmental factors [2]. Proper signal amplification is crucial when acquiring bioelectric signals, as well as minimizing artifacts resultant from environmental and biological sources. Since bioelectric signals acquisition systems are usually used in critical-care environments and in high-fidelity applications, they must fulfill a set of requirements and components.

Figure 2.9 shows a standard bioelectric signal acquisition setup, which includes signal transduction, amplification, processing and conditioning.

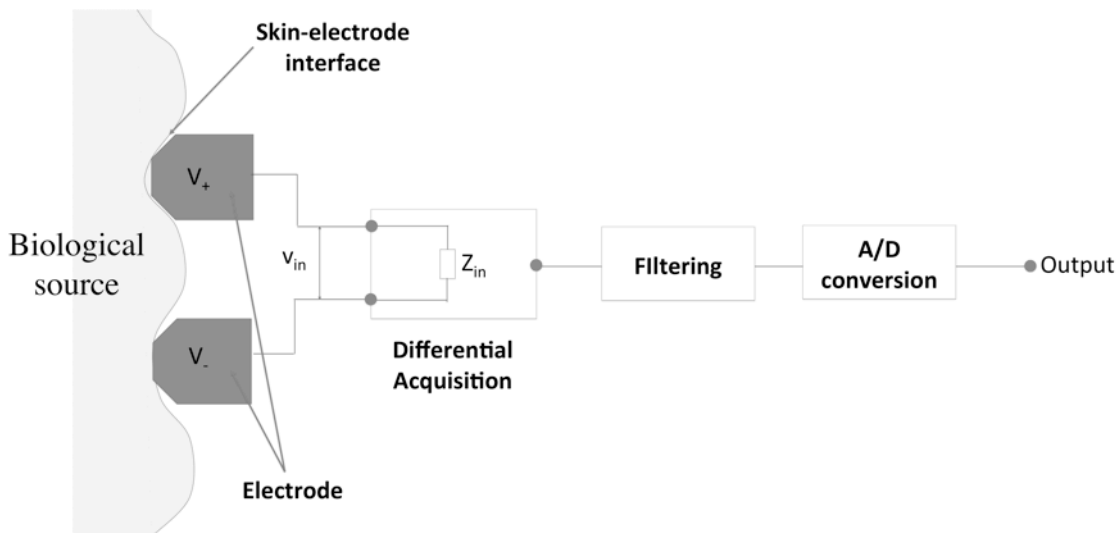


Figure 2.9 Bioelectric signal acquisition typical setup.

The differential amplifier deals with the amplification of the bioelectric signal, without compromising signal integrity. Since the input signal of the amplifier consists of the desired bioelectric signal and unwanted components (e.g. power line interference signals or other bioelectric signals), it is fundamental to include a filtering stage [25]. Generally, a notch filter

centered at 50 Hz (60 Hz in USA), and a bandpass filter are used to remove these unwanted signal components, that sometimes have higher amplitudes than the desired bioelectric signal. Finally, the setup usually includes an A/D converter to allow digital processing and communication with other units of the system and/or external devices, such as portable monitors, personal digital assistant (PDAs), among others [26].

2.2.1 Skin-electrode Interface

The charge-transfer mechanism giving origin to bioelectric acquisition takes place at the skin-electrode interface and it's of major importance in improving the design of bioelectrodes [3]. Skin-electrode interface can be modeled considering the different layers of the skin and the electrode-electrolyte interface. Generally, it is settled that the skin impedance is a combination of resistance and capacitance arranged in parallel or in series [3, 27]. This means that skin-electrode impedance is frequency dependent, and inversely related to frequency. Webster and Neuman suggested a double time constant model to describe the skin-electrode interface, as shown in Figure 2.10 [3].

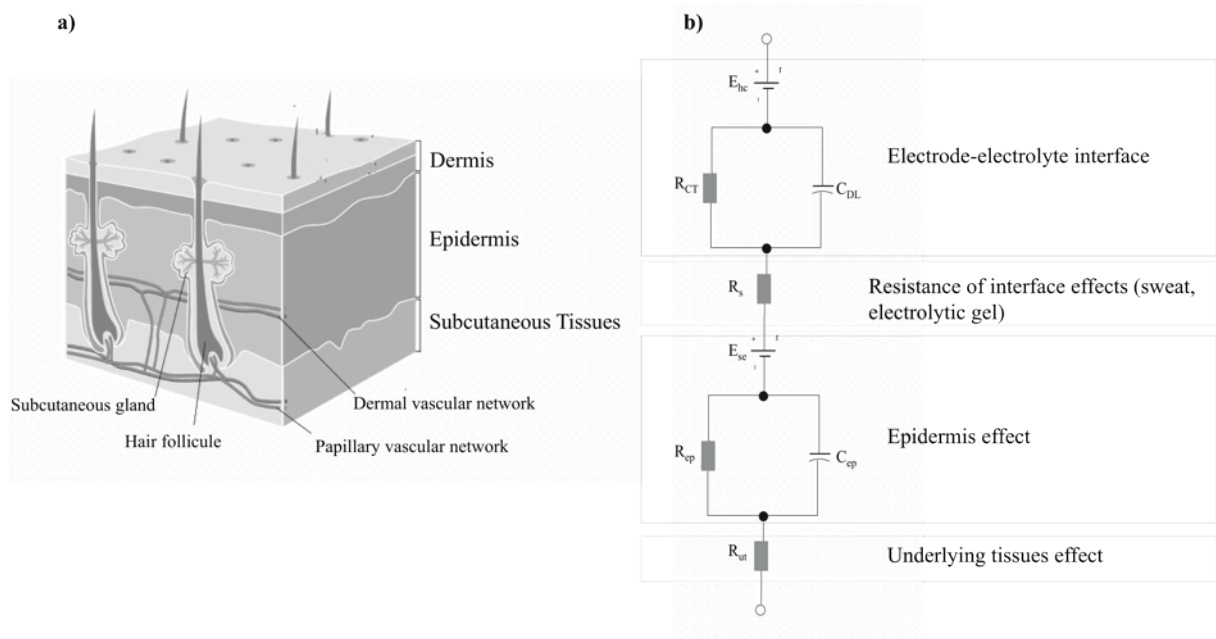


Figure 2.10 a) Human skin cross section. B) Skin-electrode interface and equivalent circuit for wet and dry electrodes.

As shown in Figure 2.10a, skin consists of three main layers: Epidermis, Dermis and Subcutaneous Layer [28]. The first corresponds to the outermost layer that is constantly renewing itself and whose role is crucial in the interface between the skin and the electrode. Also, epidermis provides a protective barrier against the hostile environment. The epidermis

is traversed by different skin additions (eg. hair follicles, sweat glands) and can be subdivided into the following layers: *stratum corneum*, *stratum granulosum* and *stratum germinativum*. The second layer of the skin, dermis, is well vascularized and contains a number of receptors for touch, temperature and pain. Dermis is composed by a dense network of connective tissue (collagen fibers), which results in higher elasticity and strength from behalf of the skin. The final layer, beneath the dermis, is called subcutaneous layer and acts a cushion to protect organs beneath the skin, as well as a fat storage [3, 28]. All layers, with the exception of the *stratum corneum*, have a rich composition of live tissue and ionic species that facilitate the conduction of electrical current [3, 4].

Figure 2.10b shows the impedance associated the electrode-electrolyte interface that includes the parallel between the reactive (C_{DL}) and resistive (R_{CT}) components. This impedance will be explained in detail in the next section. The series resistance R_s corresponds to the effective resistance associated with interface effects of the gel/sweat between the electrode and the skin. The flow of ionic current through the epidermal layer can be represented by a parallel RC circuit between C_{ep} and R_{ep} . The underlying tissues of the epidermis can be collectively represented by a pure resistance R_{ut} [3, 27]. The total impedance for the equivalent circuit is then defined as:

$$Z_T = R_{ut} + \frac{R_{ep}}{1+j\omega C_{ep}R_{ep}} + R_s + \frac{R_{CT}}{1+j\omega C_{DL}R_{CT}} \quad (2.5)$$

The electrode-skin interface could be approached by a capacitor with the *stratum corneum* forming the dielectric layer, since it stands between the electrode surface and the underlying tissues that from the second capacitor plate [29]. If so, the skin's capacitance will vary with *stratum corneum*'s thickness, dielectric constant, and electrode area, as follows:

$$C = \epsilon_r \epsilon_o \frac{A}{d} \quad (2.6)$$

where ϵ_r is the relative static permittivity, ϵ_o is the medium permittivity, A is the area and d is the distance between capacitor plates. Nevertheless, throughout this thesis, skin-electrode impedance is represented in its more discretized form as shown in Figure 2.10b.

A different model can be developed for capacitive coupled electrodes, as shown in Figure 2.11.

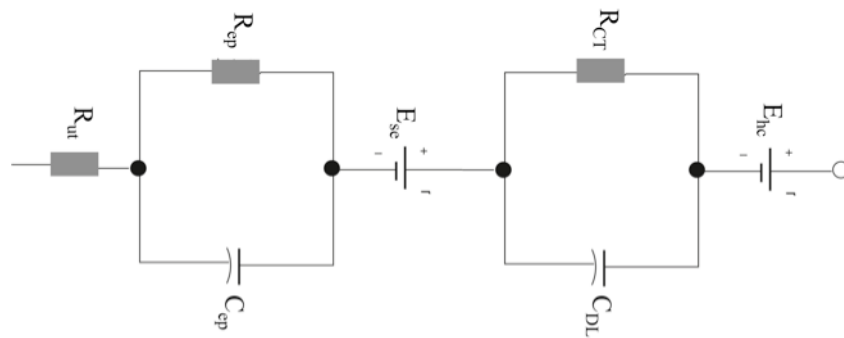


Figure 2.11 Skin-electrode interface and equivalent circuit for capacitive electrodes.

Since there isn't any electrical contact when using capacitive coupled electrodes, R_s disappears. Despite being capacitive, we consider a parallel RC circuit to electrically represent the electrodes, since we must consider always the loss component of a dielectric material. However, the frequency dependent component has the major contribution for the electrode impedance.

From the literature, we can find values for skin impedance, determined for a skin area of 1 cm^2 , from a frequency of 1 Hz to 1 MHz [30]. Figure 2.12 shows the frequency-dependent skin impedance that varied from $10\text{ k}\Omega$ to $1\text{ M}\Omega$, at 1 Hz.

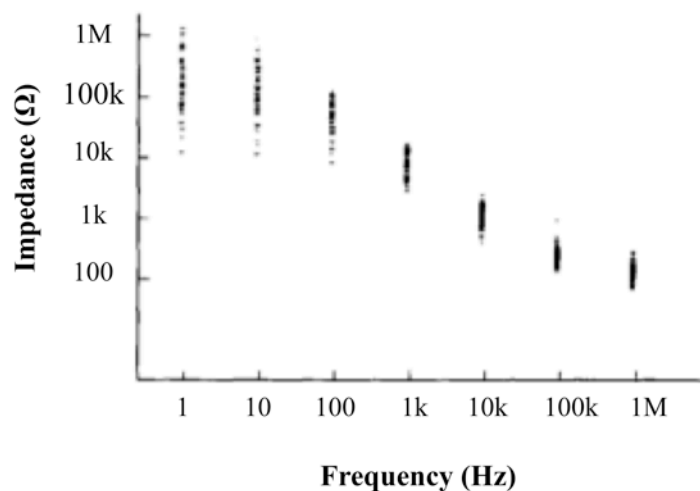


Figure 2.12 Skin-electrode impedance as a function of signal frequency [30].

Skin-electrode impedance varies with time and with recording conditions according to a set of factors, as for example: type and area of electrode, time of application, skin condition and electrolyte composition. It is recommended that skin-electrode interface for conventional wet electrodes should have an impedance below $5\text{ k}\Omega$, in order to maintain a reliable

contact [31]. To achieve this, skin is often cleaned and sometimes abraded in order to improve the stability of the bioelectric signal. However, this abrasion can sometimes be uncomfortable for the patient and even give rise to skin irritations. Considering an unprepared skin and the use of pre-gelled disposable electrodes, the reported values for skin-electrode impedance are in the order of 50–70 k Ω [32, 33]. As for the capacitive coupled electrodes, the skin-electrode impedance is usually in the range of hundreds of k Ω to a few M Ω [34].

2.2.2 Bioelectrodes

Transduction of bioelectric signals is performed by bioelectrodes, specially designed to obtain the signal of interest while reducing the potential to pick up artifact. The contact between an electrode and an electrolyte, such as in the saline environment of the human skin, results in electrochemical reactions. These are responsible for promoting the flow of electric current from the interface into the electrode wire; otherwise it would be impossible to measure a bioelectric signal with a recording apparatus [3].

The design of bioelectrodes must focus on reducing the contact impedance, improving signal acquisition while reducing the likelihood to pick up artifacts. In the past few decades, different bioelectrodes have been developed and can be classified according to material conductivity or functionality. If an electrode material is conductive, bioelectrode is classified as resistive since it establishes an electrical contact with the skin. On the other hand, capacitive electrodes are made of insulated materials that form a capacitive coupling with the skin [3, 27, 34]. Bioelectrodes can also be classified according to the type of transduction mechanism: passive (with no signal conditioning) or active (local signal processing).

Ohmic Contact Electrodes

Resistive electrodes can be subdivided into two main categories, depending on the type of interface between the electrode and the skin: wet and dry electrodes. The first type refers to electrodes that use an electrolytic gel solution to form a conductive path between the electrode and the skin. The electrolytic gel main function is to reduce skin-electrode impedance. The problem with electrodes made from electrically conductive metals as silver, copper or aluminum, resides in the fact that these are electrochemically reactive in electrolytes, and therefore, fail to provide a good pathway to electrolytic solutions or tissue. The best electrode materials are a combination of metals and their metallic salts, such as silver (Ag) in combination with a chloride coating (Cl). The result is the common and widely used Ag/AgCl bioelectric signal electrodes [3, 27].

Metal plate electrode, in its simplest form, consists on a metallic conductor in contact with the skin and an electrolyte solution. An example of this type of electrodes consists in adhesive disposable wet electrodes widely used in majority of clinical settings. Most recent metal plate electrodes are composed of a disk of plastic foam material with a silver-plated disk on the bottom surface, and a conductive lead attached to the electrodes. This attachment is made by a snap in the top surface of the plate. The electrolyte solution may be applied during the attachment procedure, or it can be already incorporated in the electrode – pre-gelled electrodes. Floating electrodes, on the other hand, have an electrolyte-insulated cavity that surrounds the metal disk, preventing interfacial instabilities due to motion artifacts [3, 4, 27].

Long-term usage of wet electrodes leads to a series of disadvantages, mainly originated from the electrolyte solution. In fact, although electrolytic solutions are effective in promoting a good skin contact, they also originate a source of noise in form of an electrical potential called skin diffusion potential [27, 29]. In addition, the reliance of an electrolyte leads to reduced signal quality due to gel dehydration, requiring reapplication of gel. Most importantly and considering a continuous monitoring for wearable applications, the application and removal of electrolytic solutions is an unpleasant and time-consuming procedure for the user and for the clinician. The use of pre-gelled electrodes can be an alternative in order to save time, but the patient would still be in contact with electrode gel that ultimately can lead to skin irritation [3, 4, 27].

Dry electrodes seek to overcome the limitations of wet electrodes, and often consist on a noncorroding metal such as stainless steel, as well as of conductive rubbers that can be repeatedly washed and reused. This metal is in direct contact with the skin and use the subject's own sweat to replace the artificial electrolyte [4, 27]. For this reason, dry electrodes tend to have better performances as perspiration accumulates in its surface, which results in a decrease in interface impedance with time [34]. Such an electrode has advantages when used in a wearable context, where patients may forget to apply electrolytic solution to the gels prior its use.

Capacitive Electrodes

Another category of bioelectrodes consists in capacitive electrodes that are characterized by the absence of electrical contact with the skin. These electrodes consist of a metal or semiconductor with a thin dielectric layer between it and the skin, which results in a capacitive coupling mechanism. When using capacitive electrodes, its surface is defined as one plate of a capacitor, and the skin is considered as the second plate [27, 34]. In addition,

there is no contact between the metal and the electrolyte, which means that in principle no half-cell potential is developed. Therefore, one source of noise during bioelectric signal acquisition is eliminated. Nevertheless, capacitive electrodes are still restricted by its intrinsic noise originated by charge accumulation and by the need for extremely high impedance readout circuits. In addition, any displacement of the electrode towards the body originates an artifact due to change of capacitance.

Usually, dry and capacitive electrodes are considered as active electrodes, whereas wet conventional transducers are called passive electrodes. In fact, the absence of electrolytic gel often implies to use active electrodes in order to transform high source impedance (skin) to a low source impedance (active electrode output). This results in the minimization of power-line hum. Other types of electrodes can be categorized, such as flexible or rigid electrodes. Flexible electrodes are the ones with adaptation ability to the inhomogeneous structure of the human skin. Examples of such electrodes are textile electrodes or other polymeric material that serves as an electrode. Novel dry and textile-based wearable electrodes have been recently proposed. These include for example conductive rubber electrodes [35], Cu sputtered textile electrode [36], conductive fabric sheets and Polyvinylidene Fluoride (PVDF) film electrodes [37], polymeric dry electrode [38].

Electrical Equivalent Model

Electrochemical reactions resultant from electrode-electrolyte interface consists in ionic solution redox, i.e. oxidation-reduction. Basically, when current flows from the electrode towards the electrolyte, oxidation occurs, being the opposite called reduction. Under equilibrium, rates of both reactions are balanced, and therefore, the current flowing in one direction is equal and cancels the current flowing in the opposite direction. Although this net current flowing is zero, due to the ion concentration fluctuations on the vicinity of the interface, a potential difference occurs known by half-cell or reversible potential [3, 27]. This potential depends on a set of parameters such as temperature, ions concentration and electrode material. The half-cell potential (E_{hc}) is particularly important in measurements involving low frequency or DC signals. Ideally, differential electrodes should have a cell potential difference of zero, i.e. their individual E_{hc} should be the same [3, 27]. However, wearable bioelectric signal electrodes are subjected to oxidation due to air exposure, staining or previous electrolyte exposure, which results in unbalanced E_{hc} . In consequence, an offset potential is added to the bioelectric signals being measured, which amplitude can reach several tens or

hundreds of millivolts. Electrode offset potentials causes current through the electrodes and through the signal conditioning circuit, being often mistaken with bioelectric potential [4, 29].

Electrical characteristics of bioelectric signal electrodes are generally nonlinear and sensitive to current density at their surface. In fact, during charge transition between the electrode and electrolyte, many must first diffuse to the interface, leading to a double layer of charge. Therefore, an interface capacitance (C_{DL}) is often included in the equivalent circuit model that characterizes the electric characteristics of the electrode-electrolyte interface (Figure 2.13a) [3, 27].

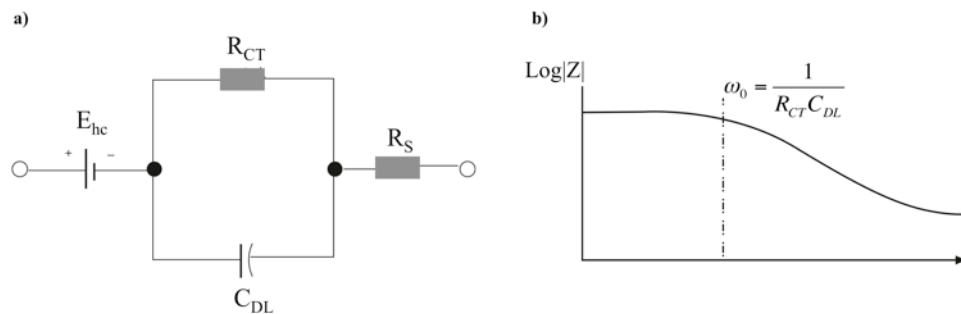


Figure 2.13 a) Equivalent circuit of bioelectric signal electrode–electrolyte interface; b) Impedance plot for equivalent circuit.

The equivalent circuit in Figure 2.13a comprises a RC parallel that represent the resistive (R_{CT}) and reactive components (C_{DL}) of the impedance associated with the electrode-electrolyte interface. The resistive component can be considered as a charge transfer resistance that shunts the nonfaradaic C_{DL} . The remaining elements correspond to the E_{hc} and a series resistance (R_s), which is essentially related with the electrolyte resistances. The electrode-electrolyte equivalent circuit demonstrates a frequency –dependent behavior, as shown in Figure 2.13b. At lower frequencies, the magnitude of the interface impedance is merely resistive since it consists on a sum of the contributions of R_s and R_{CT} . On the other hand, as frequency increases the capacitive impedance decreases whereas C_{DL} bypasses R_{CT} . Therefore, R_s dominates and equals the magnitude of the electrode-electrolyte impedance. At frequencies between these two limits, the electrode impedance is frequency dependent and thereby influenced by C_{DL} . This frequency dependency has little impact on bioelectric signal acquisition since bioelectric signals such as ECG, EEG or EMG have lower frequency components.

From an electrical outlook, a good bioelectrode should have a very low value for the resistive component, since it implies free charge transfer as well as a slight voltage drops across the interface. However, electrode-electrolyte resistance depends on several physical

properties as electrode composition, surface area and polarization. Typical skin electrodes have electrode-electrolyte resistance on the order of hundred ohms [3].

2.2.3 Bioelectric Signal Amplification

Bioelectric signal amplification is required to make it compatible with a variety of devices such as A/D converters or display equipment. These signals are recorded using a differential recording device that can be generally described as:

$$v_{BIO} = A_{diff}(v_+ - v_-), \quad (2.7)$$

where A_{diff} is the differential gain and where v_+ and v_- are the electrical potential on each of the noninverting and inverting inputs of a bioelectric signal amplifier, respectively.

A typical configuration for a bioelectric signal amplifier is called instrumentation amplifier that combines the main desirable features for this type of measurements. Instrumentation amplifiers are designed to have extremely large input impedances, high differential gain and ability to reject common signals at the differential inputs, such as power lines interference [4, 25]. This signal is often called common-mode voltage, and good instrumentation amplifier for bioelectric signal recordings requires the strong rejection of this signal. Nowadays, complete instrumentation amplifier integrated circuits (IC) are commercially available. Different considerations can be assumed depending on the type of bioelectric signal to measured. In fact, each one has a particular characteristic that makes the amplifier more prone to amplify or to remove common interference [2, 25]. Table 2.2 shows some of the special design considerations and features to take into account, during amplification stage design.

Table 2.2 Bioelectric signal-specific features and design considerations (adapted from [2]).

	Specific Features	Design considerations
ECG	mV level signal, Bandwidth (BW) of 0.05 – 150Hz.	Moderate gain, noise, CMRR, input impedance
EEG	Lower amplitude signals (microvolts)	Higher gain (>10000), low noise, higher input impedance and CMRR
EMG	Higher BW, higher amplitudes	Smaller gain, post-acquisition data processing

2.2.4 Bioelectric Signal Sensor Transfer Function

The relationship between each input of the recording device, considering the total impedance (Z_T) and the amplification stage input impedance (Z_{in}), can be described as:

$$v_{+or-} = v_{BIO+or-} \frac{Z_{in}}{Z_{in}+Z_T} \quad (2.8)$$

According to (2.8), Z_{in} of a bioelectric signal amplifier must be sufficiently high in order to avoid the attenuation of the bioelectric signal under measurement. The complete models can now be described for each approach and for the different recording situations, having in mind that the difference between them is the total impedance Z_T , which changes according to each type of electrode.

Considering the use of wet electrodes, Z_T is given by (2.5) and v_+ and v_- are easily found. For instance, for v_- :

$$v_- = v_{BIO-} \frac{Z_{in}}{Z_{in} + \left[R_{ut} + \frac{R_{ep}}{1+j\omega C_{ep} R_{ep}} + R_s + \frac{R_{CT}}{1+j\omega C_{DL} R_{CT}} \right]}. \quad (2.9)$$

The same can be done for v_+ , since a balance between the electrodes is assumed. Wet and dry electrodes are expected to have the same electrical model although the impedances values will be significantly different. In this case, R_s is related with the interface with electrode and sweat produced by the epidermal layer.

The overall bioelectric signal sensor transfer function is obtained substituting (2.9) in (2.7). As a result, for wet and dry electrodes:

$$v_{BIO} = A_{diff}(v_{BIO+} - v_{BIO-}) \frac{Z_{in}}{Z_{in} + \left[R_{ut} + \frac{R_{ep}}{1+j\omega C_{ep} R_{ep}} + R_s + \frac{R_{CT}}{1+j\omega C_{DL} R_{CT}} \right]}. \quad (2.10)$$

2.3 Wearable Bioelectric Acquisition Systems

At this point, bioelectric signals were described as well as the requirements for the acquisition of each signal. Although the essential acquisition components are similar for stationary or ambulatory monitoring, several requirements and characteristics need to be

defined for wearable applications.

2.3.1 System Components

The design of a wearable system implies three areas of work that need to be properly covered. First, it's important to develop unobtrusive wearable sensors to reliably record bioelectric data. Second, these sensors need to be implemented into a substrate material that allows for multi-sensor integration. And finally, it's important to provide infrastructures to extract and transmit data, in order to improve system performance at a clinical level and enhance mobility in individuals [39, 40].

Figure 2.14 depicts an architectural layer for an ideal wearable bioelectric system, which is composed of a functional/smart substrate, embedded electronics and attachable peripherals/appliances.

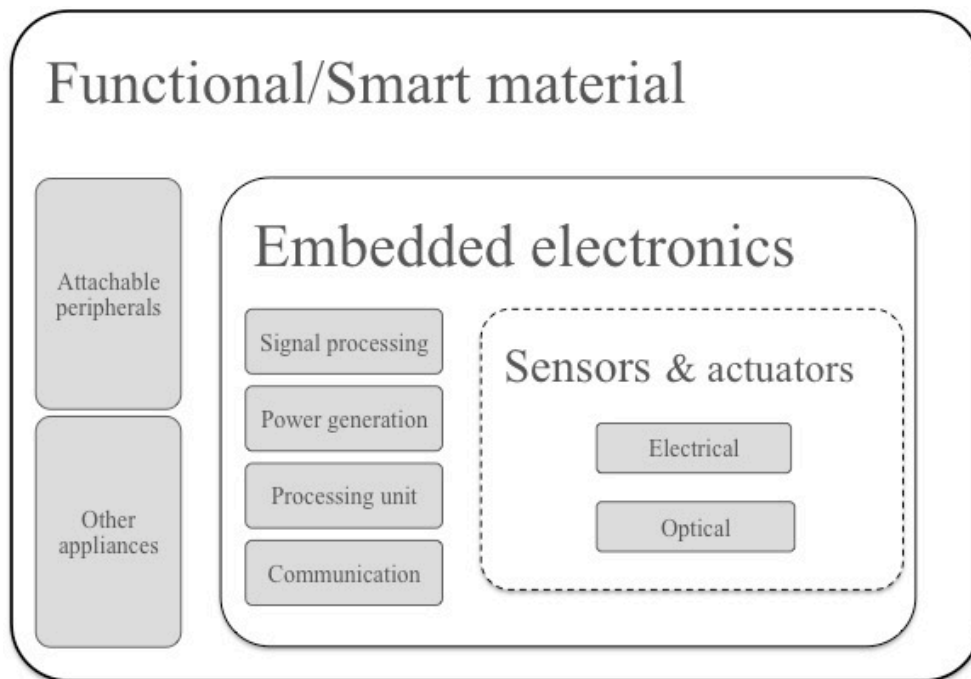


Figure 2.14 Architectural layer of an ideal wearable bioelectric system

Wearable system components may be divided into three main categories: clothing in form of a smart or functional material, embedded components, and attachable peripherals [40, 41]. The first includes all the substrate materials that act as a functional or smart structure by providing necessary supporting elements for devices that are not directly attached to the human body. Substrate materials allow the embedment of sensors, signal processing units and communication infrastructures, among others. In addition, they are responsible to provide protection from environmental conditions such as temperature changes

and humidity. The most common substrate materials used in wearable systems are textiles or flexible polymeric materials that will enable the design of normal garments with multifunctional nature [42-44]. Two approaches can be followed: one where electronic and optical components are attached into conventional clothing and accessories; or by integrating them during the manufacturing process. The latter allows creating truly functional fabrics that can be crushed and washed, whereas their properties are unaffected [40].

The embedded components include all the necessary electronics, optics, or other, that will provide sensing, actuating, signal processing, communication infrastructures, power generation, and other desired functions [40, 41]. Focusing on the sensing technologies, new approaches are required specially in providing non-contact methodologies that will improve the embedment of these components. In addition, this would contribute to the design of totally wearable and highly comfortable functional garments. Sensors are used to monitor all the necessary physiological parameters and physical environment surrounding the user, allowing to maintain the user's health condition. They can be either embedded or the material itself works as a sensing element [40, 45].

If necessary and desirable, attachable peripherals and other appliances can be included into the wearable system. Examples of these types of components are: PDAs, displays, keyboard and control knobs. Since most of peripherals are not robust enough to resist clothing-typical handling like washing and drying, they are usually associated with a particular piece of clothing or an accessory.

2.3.2 Wearability Requirements

The design of bioelectric signal acquisition for wearable devices isn't very different from regular instrumentation, despite the fact that it must fulfill the set of requirements stated in Chapter 1. Here, the main characteristics in terms of wearable systems concept and design towards maximizing the wearability will be discussed.

Wearability is classified according to a set of requirements such as low weight, small size and comfort. The main goal is to provide a device that can be carried and/or worn. The selection of the wearable material that will serve as a substrate is of extreme importance, since it determines the wearability depth, as well as the aesthetics and comfort of the device. Ideally, the substrate material used should be flexible and based on textile materials since it's possible to design systems with higher similarities to common garments. Elastic textiles or knits are the eligible materials due to their skin fitting capabilities that eventually leads to a minimization of motion artifacts and electrode displacement [46].

Location of sensors is also a major influence towards wearability, since user movements can affect the performance of the overall system. For instance, if placing the electrodes near main muscles, the muscle and movement artifact are prone to be a negative influence in the final output. In addition, the wearable system should be noninvasive, comfortable and unobtrusive, which limits the positioning of the different bioelectric sensors. The criteria selected for sensor placement depends on the functionality and accessibility needed. Nevertheless, recommended areas are those subjected to low movement and with large surface area. For instance, a sleeveless garment can be adopted for cardiac monitoring since it avoids the problems associated with limb's movements. Flexibility and applicability of the wearable system are improved if having the ability of scalability, i.e. add or remove components from the garment [46, 47].

The device should be able to interact with the environment through a network of sensors placed in different parts of the clothing or accessories. This allows to create a certain alertness of the physiological and emotional state of the user, as well as the surrounding environment. Data handling, decision support and feedback are also crucial to establish a good interaction between the device, the components and the user itself. In order to properly interact with the user, the interface should meet the principles of simplicity and friendliness, whereas minimizing the user's cognitive effort and its intervention during the process [39].

Reliability plays an important role for medical devices, especially those designed for dealing with life-threatening situations or long-term monitoring without clinical intervention. Continuous breakdowns reduce functionality of wearable devices and often lead to frustration and reduce usage on behalf of the patients.

2.3.3 Performance Requirements

Wearable system performance is driven by a set of factors related mainly with the bioelectric signal sensors used. The more general requirements are related with communication and interconnection, power supply and on-board processing. These factors are inter-dependent since the use of communication and on-board processing will increase the complexity of the system. In consequence, the power consumption will increase, affecting the autonomy of the device. Therefore, a trade-off must be established between these factors, envisioning sensor performance maximization, in terms of power autonomy.

One of the most common problems in measuring bioelectric signals is the noise and interference usually superimposed in the signal of interest. In fact, since used in a variety of situations and environments, wearable bioelectric devices are subjected to different

interference sources. Table 2.3 gives an overview of the main artifacts involved in wearable bioelectric signal acquisition systems, with indications of the peak-to-peak voltages that can be induced. It's important to determine the maximum peak-to-peak noise level acceptable mainly for ECG and EEG, since they are the most demanding signals in terms of sensitivity. Criteria selection for this threshold can be for instance considering 1% of the typical amplitudes recorded for each signal. Therefore, the acceptable noise level for ECG and EEG can be considered as 10 μV and 1 μV , respectively [29].

Table 2.3 Sources of Interference in wearable bioelectric signal recording.

Source	Magnitude fields	Frequency components
Home appliances	220 V	50 – 60 Hz
Lighting	10 kV/m	1 Hz – 1 kHz
Portable phones	1 W/m ²	>500 MHz
Microwave ovens	50 W/m ²	2.45 GHz
Skin motion artifact (stretching of the skin)	5 – 15 mV	DC
Thermal noise	0.5-10 μV	Equivalent bandwidth of the measurement device
Electrode movement	0.1 – 1000 μV	<1 Hz
Electrode-electrolyte (typical)	0.2 – 10 μV	
Skin-electrolyte interface	10 – 80 μV	

The above table shows that the most noteworthy artifacts affecting bioelectric signal acquisition is the interference from environmental sources unavoidably present in clinical or daily routine situations. In fact, since the human body is a good conductor, it acts as an antenna, coupling the electromagnetic radiation resultant from: 50/60 Hz power lines, fluorescent lighting and other equipment. The power lines interference causes intolerable noise levels in most bioelectric signals since they have components in the 50 – 60 Hz spectral band. It is very easy to have a few millivolts superimposed on the measured signal due to the power lines, which is of the same order of magnitude as the bioelectric signal itself. This interference represents a problem mainly regarding the power supply of the electronic

components used. To avoid this, batteries can be used as power supply, which eliminates the AC and DC fluctuations caused by common power line supply. However, 50/60 Hz interference may also be electromagnetically coupled to the body through electrical cables and interconnections [2, 29].

Another problem during signal acquisition is associated with subject activity, which has frequency components inside the frequency band of interest, introducing the so-called movement artifacts. These artifacts can be manifested in several forms such as skin motion artifact or electrode displacement (electrode movement, electrode-electrolyte) [27, 29]. Unbalanced effects on each electrode, also causes severe interference in bioelectric signal recordings. To eliminate this, it's important to use high input impedance measurement devices, as described in (2.10).

The overall induced body potential due to these noise and interference sources, are present at both inputs of the differential amplification stage, which can be called as common-mode potential (V_{cm}). Therefore, it is valuable to eliminate this voltage in order to prevent saturation or over-contamination of the signal of interest. In order to successfully eliminate interference or common-mode potential, it's important to design amplification systems with a high common-mode rejection ratio (CMRR) [25]. This characteristic measures the capability of the amplification system to reject interference that is equally presented at both inputs.

The overall considerations for bioelectric signal acquisition systems for wearable systems are [2, 25]:

- Supply enough gain within its bandwidth in order to reach an output level compatible with the remaining system;
- High input impedance to prevent the attenuation of the bioelectric signal, and to prevent them to be altered by other impedances variations, such as electrode impedance;
- High CMRR (> 80 dB) in order to separate as much as possible the relevant signal from noise and interferences;
- Have low output impedance and supply the amount of current necessary to the load.

Although existent technologies fulfill most of these requirements, problems associated with integration, flexibility and immunity to some interferences, such as Magnetic Resonance Imaging (MRI) rooms or others, are still a challenge.

2.4 Wearable Photonic Systems

A way to overcome the limitations imposed by electronic wearable systems mainly regarding with system integration and functionality, is the use of optical fiber-based sensors. Nowadays, optical fiber-based sensors offer the possibility of measuring other physiological signals, such as temperature, activity and blood pressure [48, 49]. In addition, optical components are already integrated in several materials and using different techniques, compatible with current textile technology [50].

2.4.1 Main Properties

Photonic technologies are based on light modulation and use optical fibers to transport it. As stated in Chapter 1, optical-based sensors have advantages when compared with electrical counterparts. An important requirement to be eligible to bioelectric signal monitoring is the ability to detect electric fields or voltages. Optical-based sensors are able to do this and to function correctly in environments where electrical interconnections fail to succeed, such as MRI rooms. Optical-based sensors are immune to electromagnetic interference, which opens the landscape of possible applications of these sensors [48, 49, 51]. In fact, it's possible to design all-optic suits with attachable power supply units in plug-in modules that can be taken off when entering in such electromagnetic interference susceptible environments.

Optical-based sensors offer the possibility of performing contactless measurements of electrical signals. This can be achieved using transducer effects by which a material exhibits an electro-optic (EO) response in the presence of a stimulus such as an external electric field (Table 1.1). By avoiding the existence of contact between the sensor and the skin, more practical and dynamic wearable solutions are available. In fact, the ideal solution is to provide the maximum comfort and flexibility to the user.

2.4.2 Main Applications

Photonic sensors are used in a variety of applications since they are able to measure different parameters such as mechanical (force, pressure, temperature), electrical and magnetic, or chemical and biological. In this thesis, the focus is towards electrical measurements. Photonic technologies are widely available for high-speed communication systems, which main areas of applications are in the military, aerospace and

telecommunication networks. All these technologies are applied to sense electric fields or to use a specific voltage signal to modulate light in order to produce EO switches.

Although photonic systems allow to eliminate the majority of electrical components and interconnections used, other optical components have to be considered. A set of requirements must be taken into account when selecting the appropriate photonic technology to use in the wearable bioelectric acquisition device. Some of these properties are shown in Table 2.4, for each type of EO modulating devices based on optical fiber [48].

Table 2.4 Photonic sensors comparison considering wearability (adapted from [48]).

	Microbending	Macro bending	Michelson Interferometer	Mach-Zehnder Interferometer	Pockels/Kerr effect
Shape	Simple	Flexible			Simple
Placement	Dependent on the final application				
Size/Weight	Small / light		Small to medium/Light to medium		-
Electronics / other optics	Simple / none		Moderately complicated / Beam Splitter, coupler	Moderately complicated / Beam Splitter, coupler, mirrors	Depends on the optical components / Polarizer
Advantages	Simple, multi-sensing	Simple, versatile	Versatile geometry, multi-sensing		-
Drawbacks	Mechanical damage of the fiber, (force) Cross-sensitivity	Mechanical damage of the fiber (bend) Cross-sensitivity	Laser needed, unknown application for low frequency signals		Unknown structure

2.4.3 Photonic Bioelectric Systems Principle

At this point, wearable bioelectric signal systems requirements were exposed, as well as the advantages of using optical-based sensors for electric field measurements. Therefore, combining the possibility of measuring electrical signals, with the wearability provided by optical-based sensors, it's possible to design systems with higher performances in a wearable bioelectric detection context.

Photonic sensors for electric field measurement operate by modulating light passing through the optical fibers, according to the effect of an external electric field [52]. This modulation can be classified according to external or intrinsic modulation, being the main difference related with their names: the use of an external device to modulate light [49].

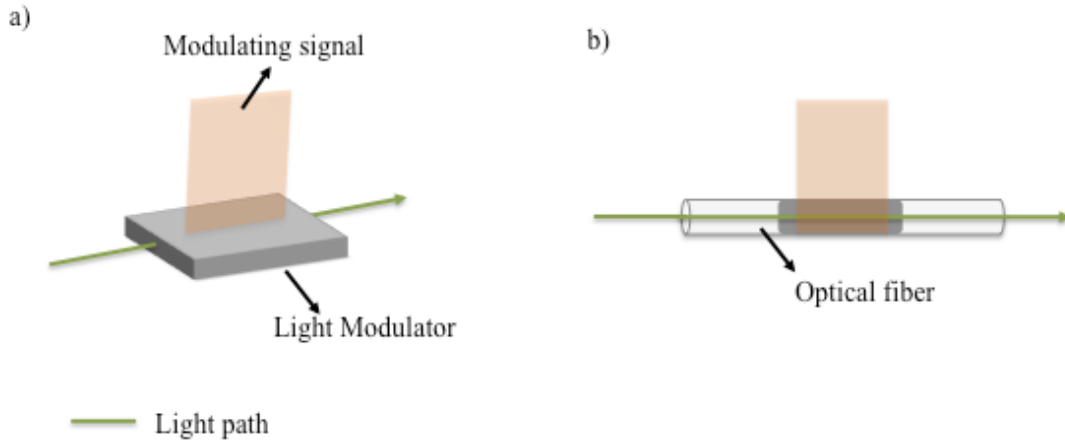


Figure 2.15 a) Extrinsic and b) Intrinsic light modulation schemes.

A modulation is called intrinsic when the optical signal source and the modulator are in the same device, i.e. optical fiber (Figure 2.15a). In this case, devices are called all-fiber sensors, and the entire fiber length is used as the sensitive area. Extrinsic or hybrid modulation consists on using the optical fiber only as a light carrier. Light is further modulated by an external optical device, as shown in Figure 2.15b. Modulated light is then carried to an optical detector. In opposition to direct modulation, extrinsic sensors performance is driven by the nature of the sensing device, instead of the optical fiber material. The main drawbacks over intrinsic modulation are the increase in production costs and complexity, as well as an increase in the overall device size. Nevertheless, extrinsic modulation is the preferred technique in this work, since it offers a higher control over the modulation [49, 53].

Depending on which property of light is modulated, modulation can be classified as intensity, phase, frequency or polarization modulation. As the name indicates, the modulation category corresponds to the light property modified by the environmental change or signal, i.e. bioelectric signal. Intensity modulation is one of the most used techniques in EO modulators, since intensity variations are more easily detected and converted to an electrical value [52, 53].

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