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# Bioimpedance plethysmography with capacitive electrodes and sole force sensors: comparative trial

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**Abstract.** Foot impedance plethysmography was implemented using two types of electrodes (dry and capacitive) and sole force sensors. The latter are commonly used for assessing diabetic foot ulcers (DFU). For impedance plethysmography, a tetrapolar configuration has been used with three different plantar setups: four skin contact electrodes, four capacitive contact electrodes and two Force Sensing Resistors (FSRs). In this work, FSRs have been considered as possible capacitive electrodes because the top substrate contains interdigitating conductive electrodes and a semiconductive polymer. All the measurements have been performed using a 1 mA/10 kHz excitation current and have been tried under the feet of a standing person to detect impedance plethysmography signals. Contact electrodes allow a good cardiac pulse signal while capacitive contact through the socks features mains interferences. Force sensing resistors with their force-dependent resistance in parallel to the capacitive coupling, were not able to detect cardiac pulse. But promising results can be anticipated from these findings provided higher frequencies are used and larger sensor areas to help detect altered skin states in diabetic foot.

**Key words:** Force sensing resistor, capacitive coupling, impedance plethysmography, bioimpedance

## 1. Introduction

Electrical bioimpedance is a characteristic of living tissue which is determined by stimulating with currents of known amplitude and frequency. Living tissues, suspensions and biological matter can be considered linear for the purpose of characterization and recognition. As research progresses, applications are being developed to monitor physiological processes and biochemical states at very low cost and in the absence of any known adverse effect.

Often a reasonable alternative to costly and hazardous procedures such as nuclear medicine, electrical bioimpedance fulfills clinical monitoring needs [1]. One such clinical objective is to monitor epidermal foot state during gait, and specifically for diabetic patients at risk of serious foot lesions [2]. The functional integrity of the skin is reflected in its electrical impedance spectrum and can be used to help manage diabetic foot ulcer (DFU) [3].

Wearable devices are being extensively developed and will be used provided they do not interfere with normal life. A major contribution to persons with diabetes would be the prevention of DFU, not only in general terms, but specifically by stopping the preliminary pathological processes in time before any actual DFU is actually installed [2]. Several physical variables (multifunctional sensing [4]) are fueling the development of new wearables. Comfort is a high key performance indicator to reach patient



adherence to any wearable device. Sensors such as textile electrodes or dry electrodes (stainless steel, carbon rubber or metallized fabric) [5] improve device usability and form factor and, in terms of long term monitoring, reduce skin contact adverse effects.

Present day gold standard for DFUs follow-up is plantar pressure [6] and foot microclimate. Additionally, foot status can be better mapped with plantar blood flow under the first metatarsal head [7] and with electrical bioimpedance data [8]. The concept of multiple variables to be monitored during gait includes the existence of several sensors around the foot. Force Sensing Resistors (FSR) are commonly used in pressure sensing applications because of their shape, size, and flexibility. These applications usually involve high peak pressure points to detect critical areas or possible DFUs, but lack the dynamic components of gait [9].

Impedance plethysmography (IPG) is a bioimpedance measurement technique that allows us to obtain arterial volume variations of the arterial circulation [10]. IPG could be an additional measurement to be included in multivariable approaches to the monitoring of diabetic foot during gait as it adds cardiac activity. Thus, the present study aims to understand the behavior of a pair of FSRs used as impedance electrodes to obtain cardiac activity and eventually foot composition. Bioimpedance is usually measured with either skin contacts or contactless electrodes. Since the diabetic foot monitoring includes pressure sensors, our aim in the present study is to check the possibility to use the FSRs both as pressure sensors and bioimpedance electrodes. In a dynamic scenario, FSRs could measure pressure during the stance phase and cardiac pulse during the swing phase of gait [11].

## 2. Materials and methods

To ascertain the possibilities of using plantar pressure sensors FSR, with their metallic parts, as bioimpedance electrodes, we have designed three set-ups with associated acquisition and processing, of an impedance plethysmograph, as shown in Figure 1 [10] and [12]. A 1mA/10 kHz current is generated by a Voltage-Controlled Current Source (VCCS) and is injected by two electrodes, one in each foot. The other two electrodes are connected to the amplifier described in Figure 2 to measure the voltage response.

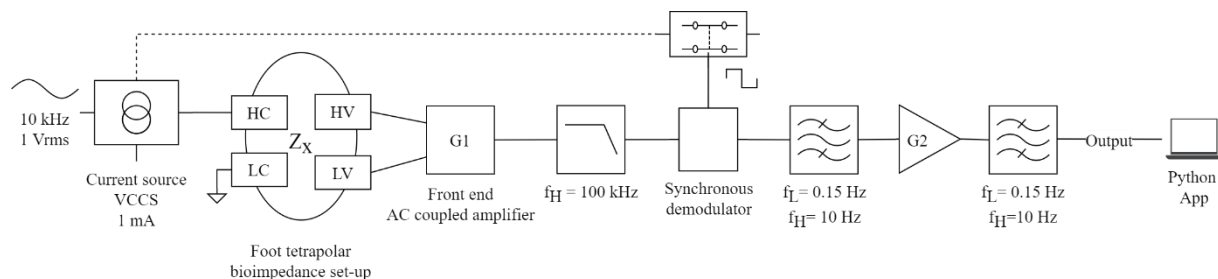


Figure 1. Foot plethysmography with four bioimpedance electrodes. Current is injected by means of HC and LC electrodes; the resulting voltage is measured across HV and LV.

### 2.1. Capacitive electrodes

Nowadays, biological data acquisition tends to be as non-invasive as much as possible to maintain high signal quality. Capacitive methods allow to measure (through thin fabric or coating) relevant physiological information such as biopotentials and electrical bioimpedance.

The interface of electrodes that are not in contact with the skin is more complex compared to conventional electrode-electrolyte-skin electrodes. For non-direct measurement, the capacitive electrode will act on an ideally thin surface or coating. This coating will prevent the electrode from coming into direct contact with the skin. So, there is no galvanic current in capacitive set-ups [13].

### 2.2. Force sensing resistors

A FSR has a variable resistance depending on the pressure applied to the active area. FSRs consist of thin polymer and ultem layers (polyetheramide, semi-transparent yellow substrate, with chemical resistance and excellent temperature and limited flexibility) that decrease in resistance with increase in

force applied to the active surface. A FSR is not a cell or strain gauge, however, it has similar properties [14].

The device is made of elastic materials in four layers. The first layer is a plastic electrical protection; the second layer is the active area of 18.3mm in diameter, consisting of a conductor pattern, which is connected to the tip of the tail to give an electrical voltage; a plastic division allows air ventilation through the tail; finally, a layer of flexible substrate covered with a thin conductive layer of polymer and ultem is aligned with the active area. Also, the device has a sensitivity range of force from 100 g to 10 kg, a resting resistance greater than 1 MΩ and maximum force resistance of approximately 100 kΩ; life time of 10 million activations [14].

The feasibility of using FSR pressure sensors as bioimpedance plantar electrodes relies on the four layers composition of the sensor. First and third layers work as insulators, second layer as top substrate and the fourth layer as conductive bottom substrate. The top substrate contains interdigitating conductive electrodes and the bottom a semiconductive polymer [15]. Thus, it could operate as a capacitive electrode.

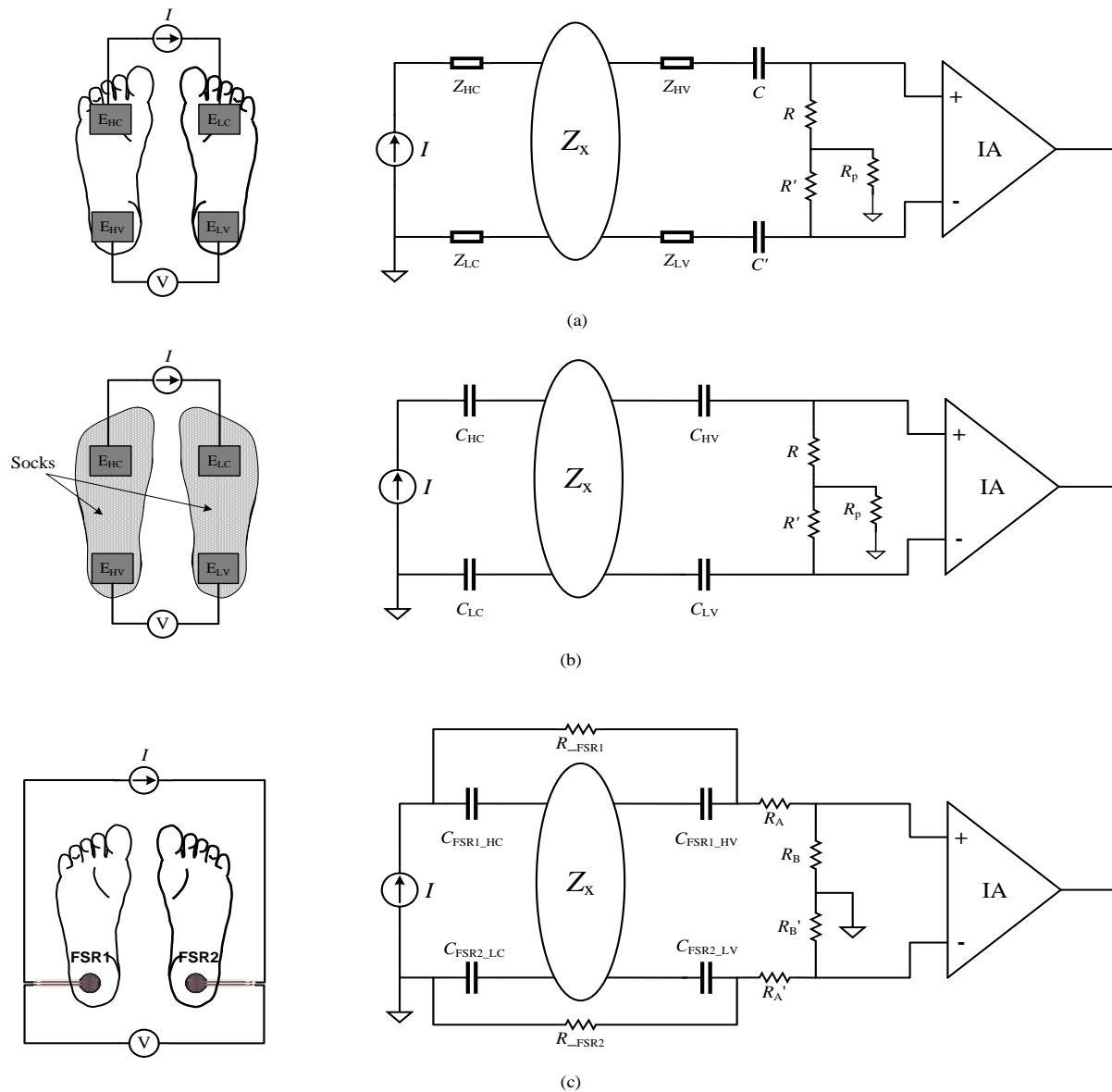


Figure 2. Tetrapolar plantar set-ups and their equivalent circuits. a) skin contact electrodes b) capacitive electrodes across socks c) skin contact FSR sensors.  $Z_x$  represents foot to foot bioimpedance.

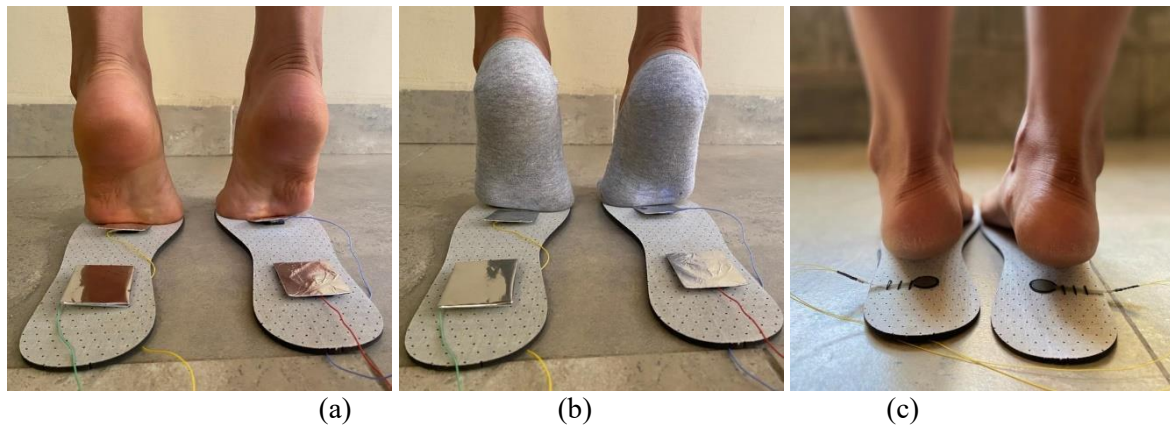


Figure 3. Experimental set-ups. a) bare feet on soles with skin contact electrodes aluminum pads b) capacitive electrodes across socks and c) bare feet on soles with skin contact FSR sensors.

### 2.3. Tetrapolar two feet heel skin contact set-up

Figure 2 shows the equivalent circuits of tetrapolar configurations for three measurement setups. To measure the voltage, an alternating-coupled front end suppresses the contribution of mains interference and the contribution of  $1/f$  noise from the instrumentation amplifier (IA).

Figure 2a shows a configuration used in [7] and [10], applied in electronic weighing scales that have dry electrodes to perform Bioimpedance Analysis (BIA). For this set-up, Al square plates (4 cm x 4 cm) were used as skin contact electrodes, soldered to 1 mm<sup>2</sup> section insulated wires. Two electrodes were located under each heel of a healthy subject (one of the authors).

### 2.4. Tetrapolar two feet heel capacitive set-up

The same materials, circuits, and configuration of Figure 2a have been used but, in this case the electrodes are located outside and under the socks, which transforms the electrode coupling in a capacitive coupling. In Figure 2b, a new configuration is presented, it is suggested to dispense direct contact within the foot sole, taking measurements across the socks. In this configuration, the contact capacitances  $C_{HC}$ ,  $C_{LC}$ ,  $C_{HV}$  and  $C_{LV}$  depend on the fabric of the sock and the area of the electrode plate.  $C_{HV}$  and  $C_{LV}$  form a differential high-pass filter with the bias resistors  $R$ ,  $R'$  and  $R_p$ , thus eliminating the contribution of line mains interference and the contribution of  $1/f$  noise from the IA. In this sense, the values of  $R$ ,  $R'$  and  $R_p$  must be chosen to guarantee a frequency response that minimizes the gain error of the IA at the excitation frequency. Following Luna and Pallas [16], who use capacitive electrodes in a chair, in this work it is suggested to perform plantar bioimpedance measurements using capacitive electrodes.

### 2.5. Tetrapolar two feet heel FSR capacitive set-up

In Figure 2c, the equivalent circuit using two FSRs as electrodes is shown. With this configuration, the extension of each contact terminal of the FSR is assumed as a capacitive contact ( $C_{FSR}$ ) with the subject's skin, which is due to the insulating film that exists between the active area of the sensor and the foot. In addition, between both contact terminals, there is also the resistance ( $R_{FSR}$ ), which depends on the force exerted on the active area of the sensor. The greater the force exerted, the lower the value of  $R_{FSR}$ . Unlike the configurations of Figures 2a and 2b, a current path between the current injection and voltage sensing capacitances is formed through  $R_{FSR}$ , that is, bypassing the impedance of interest  $Z_X$ . In this sense, due to the  $C_{HC}$  and  $C_{LC}$  capacitances (30 pF, according to preliminary tests), the amount of current that flows through  $Z_X$  will depend on the frequency of the injected current. That is, at low frequencies, a smaller amount of current would flow through  $Z_X$ , while, as the frequency of the injection current increases, more current will flow through  $Z_X$ . The residual current finds a ground path through  $R_{FSR}$ , and the inputs of the IA. If a configuration like that presented in Figures 2a and 2b were used, the voltage seen at the

inputs of the IA would saturate it. That is why, in Figure 2c a voltage divider formed by  $R_A$ - $R_B$  and  $R_A'$ - $R_B'$  has been arranged, where  $R_A > R_B$  ( $R_A' > R_B'$ ), to attenuate the input voltage and avoiding the saturation of the front-end. Nonetheless, this is a solution that contrasts with the typical solutions of a voltage measurement to reduce loading errors, but in this case, it may be considered (a priori) a simple and optimal solution.

### 3. Results

A tetrapolar bioimpedance signal stimulated with 1mA/10 kHz current using skin contact electrodes, capacitive electrodes and two FSRs are shown in Figure 4a, Figure 4b and Figure 4c, respectively. Figure 4a and 4b show the cardiac signal obtained from IPG measured from foot to foot. The signal with dry electrodes and the signal with capacitive electrodes both shows cardiac activity in a remarkable way. In Figure 4b, the large 50 Hz interference is due to the capacitive coupling to  $Z_X$ .

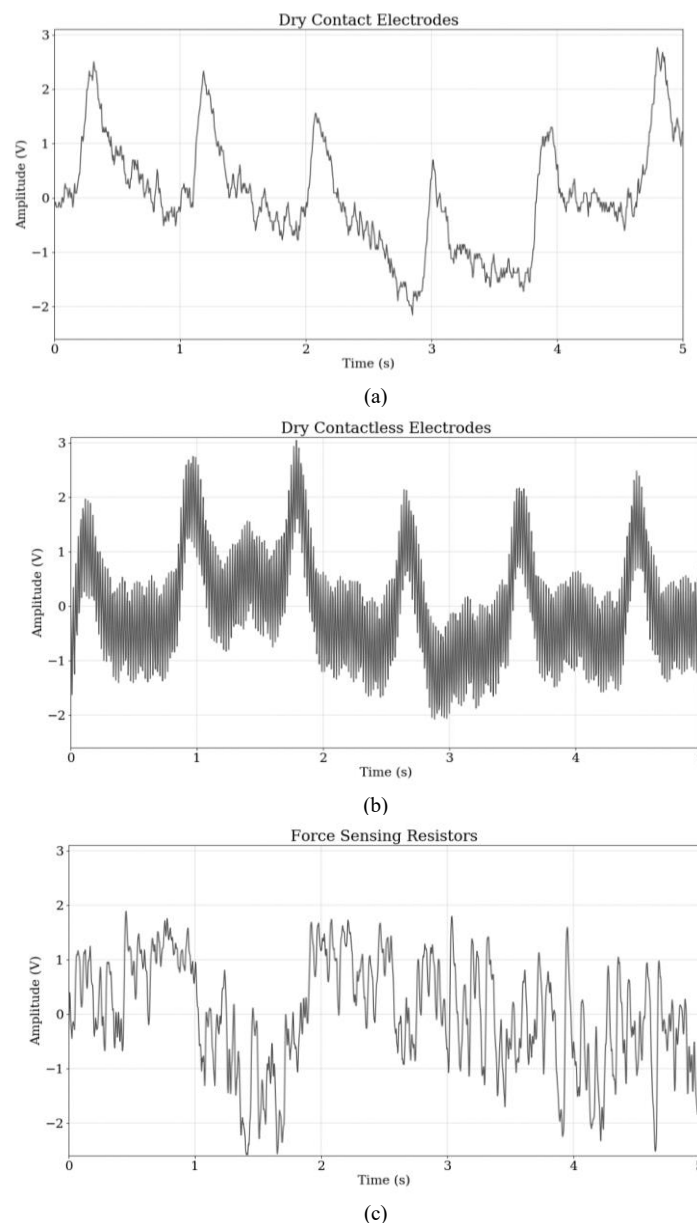


Figure 4. Signals obtained from foot-to-foot Impedance Plethysmography corresponding to Figure 2: a) skin contact electrodes b) capacitive electrodes across socks c) skin contact FSR sensor.



In Figure 4c, when using FSRs as electrodes, the cardiac signal is not detected. To explain these results, several hypotheses are considered: first, the stimulating frequency (10 kHz) causes the injection capacitances  $C_{HC}$  and  $C_{LC}$  offer a high impedance to the injected current. As a result,  $I$  flows mostly through  $R_{FSR1}$  and  $R_{FSR2}$ , which are of the order of hundreds of ohms because of the mass of the subject on the FSR. Second, the contact area of the "electrodes" may be too small to sense impedance changes due to arterial volume variations. Finally, muscle and motion artifacts alter the  $R_{FSR}$  values at random, a phenomenon that can bury the cardiac activity in noise.

#### 4. Discussion and conclusion

In the quest for variables to be monitored on the skin of diabetic foot persons, we have tried to implement impedance plethysmography measurements with dedicated multiple electrodes. The electrodes consist of 4x4 cm aluminum pads, while multifunction was investigated for FSR sensors, which are presently recording pressure during the stance phase of gait. The contact of the aluminum pads was either directly to the skin or through the socks: either dry resistive or capacitive.

Normal IPG signals were obtained with dedicated conventional electrodes (4x4 cm aluminum pads), resistive and capacitive contact, but with multifunction FSR contact, no cardiac activity was detected even though a different front-end configuration was implemented. Thus, higher frequency averaging or other techniques should be further investigated.

The main problem with the measurement using FSRs is that a considerable current bypasses  $Z_x$  through  $R_{FSR}$ , which reduces the availability of a large enough signal to be detected. To minimize the bypass, there are two possibilities: i) to increase the excitation frequency to reduce the capacitive reactance of  $C_{HC}$  y  $C_{LC}$  or ii) to increase the FSR resistance by relieving the weight of the person standing on the force sensor. Higher resistance due to lower weight can be accomplished in sitting or supine positions, albeit this option can only be considered as a calibration instance or during the swing phase of gait.

This study is the starting point to develop multimodal measurement systems using the smallest number of sensors in subjects with diabetic foot. Although the feasibility of obtaining IPG with no direct contact with the skin using metallic electrodes was demonstrated, the use of FSR proved to be more challenging. Therefore, further work is required to adapt FSR as successful capacitive electrodes to detect cardiac activity.

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