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Characterization and comparison by bioelectrical impedance analysis of dry electrodes for use on human quadriceps, in a connected garment

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Abstract

Muscle tissue exhibits strong anisotropy due to the orientation of its fibers, and its electrical properties vary during contraction. Consequently, local impedance measurements can be used to detect muscle activity. This approach is particularly relevant for rehabilitation monitoring and serious game applications, where real-time physiological feedback is required.

The objective of this study is to characterize and compare a set of dry electrodes of different types (textile and silicone), first on a phantom simulating human quadriceps tissue, and subsequently on two human subjects. The goal is to identify a suitable electrode for integration into a real-time connected garment.

For each electrode, bipolar and quadripolar measurements of resistance (R) and reactance (X) were performed, and impedance (Z) was calculated. The electrodes were initially characterized on a phantom and then tested on the quadriceps of two adult subjects (one male and one female). The results show that, although the impedance values obtained are consistent

with those reported in the literature for quadriceps tissue, they differ significantly from those obtained with the reference electrode.

Furthermore, a significant variability was observed in quadripolar measurements depending on the electrode type. Nevertheless, this study demonstrates that electrodes can be selected based on their electrical properties and integrated into a connected garment, with performance comparable to that of standard gel electrodes.

Keywords: phantom; bioimpedance; agar-agar; quadriceps; muscle; dry electrodes

1. Introduction

According to its contemporary definition, a *Serious Game* refers to a "playful simulation designed for serious purposes such as learning." The concept is believed to have originated in the early 19th century, initially as a method for military training. In the late 20th century, following the publication of Abt's seminal work [1], the scope of Serious Games expanded beyond military applications to include domains such as education, politics, and marketing. These applications can take various forms, including board games, outdoor activities, and digital or computer-based environments.

Since the early 2000s, growing interest in Serious Games has led to numerous academic studies, the establishment of classification frameworks, and the development of dedicated databases by Alvarez, Rampnoux, and Djaouti [2–8]. More recently, these approaches have been increasingly adopted in the healthcare sector, where they aim to enhance patient engagement by introducing interactive and motivating elements into medical environments. In clinical settings, Serious Games can serve multiple purposes, such as preparing patients for surgery, supporting chronic disease management, facilitating rehabilitation, and training healthcare professionals.

This period has also been marked by significant technological advances, particularly in information technology and the emergence of virtual reality (VR). The first VR headset was developed by Daniel Vickers in 1970 at the University of Utah, and subsequent efforts by both research institutions and industry have progressively improved the accessibility of this technology (e.g., VPL Research, Sega, Apple, Google) [9–13]. The biomedical field has leveraged these innovations in computing, communication technologies (notably smartphones), and electronics (reduced cost, improved performance, and miniaturization) to develop new health and wellness solutions. In particular, there has been a growing interest in wearable systems incorporating embedded sensors, such as ECG devices and smart garments, which rely on reusable, washable dry electrodes that do not require conductive gels [14–15].

These biomedical monitoring systems generate large volumes of data, often referred to as Big Data. Their effectiveness depends on the ability to account for individual physiological characteristics during rehabilitation, thereby enabling personalized adaptation of therapeutic or training protocols. In this context, the present work is part of a broader project involving Serious Games based on virtual and augmented reality, designed for healthcare professionals, athletes, and the general population, and relying on an ecosystem of electrophysiological sensors.

Muscle contraction induces variations in the electrical properties of muscle tissue [16–18]. These variations can be detected through local impedance measurements, enabling the identification and monitoring of muscle activity. In rehabilitation and Serious Game applications, real-time tracking of muscle contraction is particularly valuable, as it allows dynamic adjustment of exercise intensity and progression based on physiological parameters such as muscular fatigue. Bioelectrical impedance analysis (BIA) constitutes a central component of this approach, although additional physiological signals may be considered in future developments depending on the intended application.

A key challenge in the design of smart garments lies in the selection of appropriate electrodes. While gel electrodes are commonly used for bioimpedance measurements, they are not well suited for wearable applications due to their limited reusability and lack of washability. Consequently, alternative electrode technologies must be considered. In the context of connected garments, electrodes must be washable, reusable, and capable of ensuring reliable and accurate acquisition of bioelectrical signals [19–21]. Although a variety of textile electrodes are currently available, their performance in BIA applications remains insufficiently characterized. Therefore, a systematic evaluation and comparison of these electrodes is required, initially on a phantom model and subsequently on human subjects [22–23].

The electrical properties associated with the electrode–tissue interface are assessed using bipolar measurements, whereas the intrinsic properties of the phantom or muscle tissue are obtained through quadripolar measurements. The analysis is based on the concept of bioimpedance, denoted as Z, and defined as:

$Z=\sqrt{R^2+X^2}$Z=R2+X2

where R represents resistance and X represents reactance.

For quadripolar measurements, the relationship between reactance and resistance is analyzed using the Cole–Cole model [24], which enables the characterization of the electrical behavior of biological tissues based on impedance data (Figure 1).

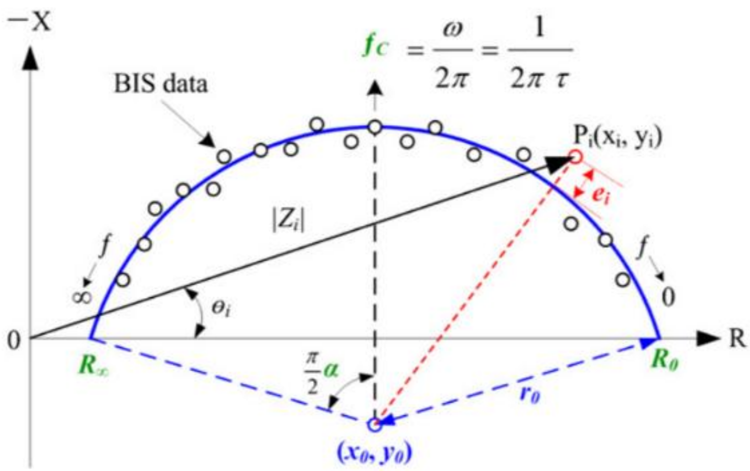


Figure 1. Cole-Cole curve model [24]

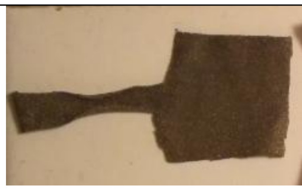
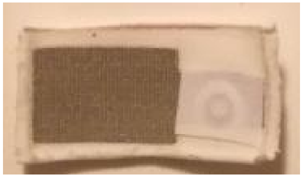
The primary objective of this study is to characterize a set of dry electrodes of different types (textile and silicone), first on a phantom simulating human quadriceps tissue and subsequently on two human subjects. The electrodes are initially evaluated using an agar–agar gel phantom developed in a previous study [25]. They are then tested on a small population (n = 2), which served as a reference for the phantom design. Their performance is compared with that of conventional gel electrodes commonly used for muscle activity detection. Ultimately, this work aims to identify a suitable electrode for integration into a real-time connected garment prototype.

2. Material and Methods

2.1. Material

Table 1 describe the tested electrodes:

Table 1. Description of the tested electrodes. 3M RedtDot electrodes are considered as our reference

Electrodes	**Dimension**	**Références**	**Photo**
Statex	32x32 mm Area = 1024 mm²	Shieldex Technik-Tex P130-B	
Sol Gel	46x22 mm Area = 1012 mm²	Prototyped electrodes based on the same Statex Textile with a dedicated coating (Carroucell SAS)	
Knitted Noble	29x19 mm Area = 551 mm²	Knitted electrodes using Noble wire	
Conductive Silicon	32x32 mm Area = 1024mm²	sold by the meter for electrotherapy (https://www.kipocora.fr)	
Repositionable monitoring electrodes 3M Red Dot electrodes Ag/AgCl gel	32x32 mm Area = 1024mm²	3M gel electrode	
Repositionable EMG electrodes Spes Medica	32x32 mm Area = 1024mm²	TEON1F090 based on adhesive conductive silicone	

The intrinsic resistance (ohm/mm) and the anisotropic character (0/90°) of the electrodes are determined with a multimeter.

Impedance measurements (Resistance R, Reactance X and Impedance Z) are performed with a bioimpedance electronic board (ZRange (0–2500 ohms); Zreal Coefficient of Variability

(CV) % 0.025% and Zimg CV% 0.408%; accuracy (all frequencies) Zr mean error 2.60% ± 2.01% and Zimg mean error 0.43% ± 0.84%) (Figure 2).

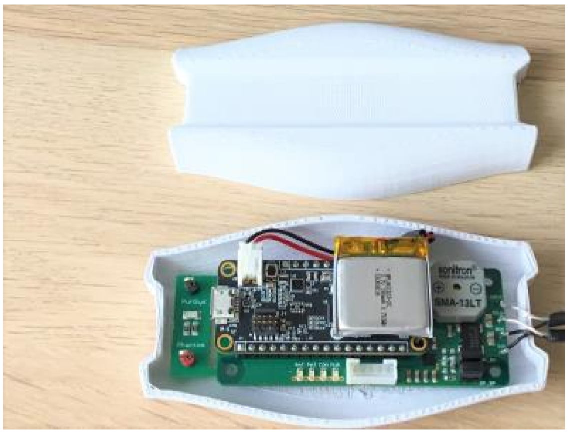


Fig 2. Electronic board of bioimpedance BioZ (Module Phi-Light, RunSys, France)

The electrodes are characterized on an AA phantom (Figure3) with electrical properties like a human quadriceps (at 4kHz R=67.1 ohm and X=7.57 ohm) [17]. The AA phantom is made according to the protocol established in a previous study [26]: agar concentration of 4g/100ml of demineralized water, NaCl of 7g/L and graphite of 6g/100ml.

Fig 3. AA phantom simulating a human quadriceps (NaCl concentration = 7g/L; graphite concentration = 6g/100ml).

The temperature and hygrometry of the environment are observed during all measurements.

For intrinsic measurements, the electrodes are placed directly on an insulating wooden support.

2.2. Characterization of intrinsic material properties

The electrodes were first characterized intrinsically in order to assess their material properties. The electrical resistance of each electrode was measured in the absence of injected current using a multimeter. Measurements were performed at orientations of 0° and 90° to evaluate the anisotropic behavior of the materials without current injection. The 0° orientation corresponds to the fiber direction (for silicone electrodes, 0° corresponds to the direction perpendicular to the manufacturing line, referred to as the "banana line"). For each electrode and each orientation, four measurements were conducted to ensure repeatability.

Since the presence of an electric current influences electrode behavior [27], additional characterization was performed using bioelectrical impedance measurements. The effects of orientation and frequency were investigated following a protocol similar to that used for measurements without current injection. The impedance ZZ was calculated according to Equation (1). To assess the influence of frequency on the measurements, experiments were carried out at six frequencies: 4, 8, 18, 40, 80, and 128 kHz.

As the electrodes differ in size and geometry, the measured values of resistance (R), reactance (X), and impedance (Z) were normalized by the distance between the two measurement points of the multimeter or impedance module, corresponding to the diagonal length of each electrode. This normalization allows comparison of the electrical properties over an equivalent measurement distance.

2.3. Characterization of electrodes on AA phantom

Prior to testing on human quadriceps, the electrodes were first characterized using an agar–agar (AA) phantom with electrical properties similar to those of human quadriceps tissue [25]. The gel electrodes (3M and Spes Medica) were systematically tested last in order to avoid possible gel residues that could influence the measurements performed with the textile and silicone electrodes.

The electrodes were arranged in linear configurations, and a pressure of 0.26 g/cm²—previously determined in an earlier study [25]—was applied to ensure a stable and reproducible contact interface (Figure 4).

Fig. 4 Mounting to progressively add weight to the AA phantom

To ensure thermal equilibrium, the phantom was removed from the refrigerator one hour prior to the measurements and allowed to reach room temperature. For each electrode, five bipolar and quadripolar measurements of resistance (R) and reactance (X) were performed, and impedance (Z) was calculated using Equation (1). Bipolar measurements were used to characterize the electrode–phantom contact interface, whereas quadripolar measurements reflected the intrinsic electrical properties of the simulated quadriceps tissue.

As for the intrinsic characterization, the values of R, X, and Z were normalized to allow comparison between electrodes of different geometries. For bipolar measurements, normalization was performed with respect to the electrode surface area, while for quadripolar measurements, normalization was based on the volume through which the current flows. This volume was approximated by a half-torus geometry and calculated according to Equation (2):

$$V=\frac{n}{4}(D-d)^2\times(D+d) \quad (2)$$

Cole–Cole diagrams were then plotted for each electrode to analyze their behavior under current injection. The impedance values obtained were compared with those of the reference electrode (3M).

When electrodes are applied to human skin, their behavior evolves due to factors such as body temperature and perspiration before reaching a stable state. Therefore, a stabilization period prior to measurement must be defined [28]. For this purpose, the electrodes were positioned linearly on the subject, with a distance of 16 cm between the centers of the two inner electrodes (Figure 5), following the configuration proposed by Nescolarde [18].

Immediately upon contact with the skin, low-frequency measurements (4 kHz) of R and X were initiated using the bioimpedance module. Data acquisition was performed over a duration of 5 minutes to monitor stabilization [28]. During this phase, the electrodes were maintained in contact with the skin without applying additional pressure, using an insulating material (household silicone gloves).

The study on human quadriceps was conducted on two healthy adult subjects (one male and one female). All participants provided informed consent prior to participation. The experimental protocol was reviewed and approved by the Human Use and Ethics Committee (CEUA) of ENSIIE (Paris, Protocol_phase 1, 2020) and was conducted in accordance with the principles of the Declaration of Helsinki (World Medical Association, June 1964).

For the experimental measurements, the electrodes were positioned on the quadriceps following the same configuration as described by Nescolarde [18]. They were secured using an elastic adhesive band (elastoplast-type). Subjects were instructed to adjust the applied pressure to match that of a tight-fitting sports garment (e.g., leggings). Measurements were then performed after the stabilization time previously determined.

Fig. 5 Positioning and maintaining the electrodes on the subject's quadriceps

2.4. Data Analyse

Each dataset was first tested for normality using the Shapiro–Wilk test in order to determine whether parametric or non-parametric statistical methods should be applied.

The effects of frequency, electrode orientation, and electrode type were assessed using the non-parametric Kruskal–Wallis test. Pairwise comparisons between the values obtained with each type of electrode were performed using the Mann–Whitney test.

All statistical analyses were conducted using XLSTAT software.

3. Results

3.1. Characterization of material properties - intrinsic electrode measurements

3.1.1. Intrinsic resistance Ri measured with a multimeter (ohm/mm)

Table. 2 Coefficient of variation in % of R depending on the electrode and the positioning angle

Electrodes Type	**Angles**	**Variation Coefficient %**
Ag/AgCl Ref 3M	0	Reference
Knitted Noble	0	34.9
Textile Statex	0	43.5
SolGel Dry	0	49.6

Silicon	0	14.9
Reusable Spes Medica	0	2.25
Ag/AgCl Ref 3M	90	Reference
Knitted Noble	90	16.5
Textile Statex	90	13.4
SolGel Dry	90	34.8
Silicon	90	11.5
Reusable Spes Medica	90	13.7

A high coefficient of variation (%) is observed (Table 2), which is likely due to the limited contact area between the measurement probes and the electrodes. This variability may also be attributed to the structural characteristics of the electrodes. In particular, the Knitted Noble electrode exhibits gaps between the yarns; depending on whether the probe is positioned on a yarn or within a gap, significant variations may occur.

For the SolGel electrodes, in addition to the presence of inter-yarn gaps, the heterogeneity resulting from the mixture of titanium-based elements and carbon may further contribute to variations in intrinsic resistance, especially as a function of orientation. This effect is particularly pronounced at 0°. Overall, all textile electrodes (Knitted Noble, Statex, and SolGel) exhibit relatively high variability, ranging from approximately 30% to nearly 50%.

However, intrinsic resistance does not necessarily reflect the behavior of the electrodes under current injection. When considering the coefficient of variation obtained from impedance measurements, the effect of orientation appears to be statistically significant ($p = 0.0002$; $p < 0.01$), highlighting the influence of material anisotropy on conduction properties (Table 3). This factor should therefore be carefully considered during electrode integration and assembly.

Table.3 P-value obtained for the different electrodes compared to the reference

Electrodes Type	**p-value**
Ag/AgCl Ref 3M	Reference
Knitted Noble	0.518
Textile Statex	0.099

SolGel Dry	0.518
Statex	0.099
Silicon	0.099
Reusable Spes Medica	0.099

However, no significant difference in intrinsic resistance was observed between the different electrodes and the reference electrode.

3.1.2. Bipolar bioimpedance characteristic of the electrodes (ohm/mm)

Table. 4 Coefficient of variation in % of Z depending on the electrode and the positioning angle

Electrode type	**Angles °**	**CV Z (Ohm.mm)**
Ag/AgCl Ref 3M	0	0.004
Ag/AgCl Ref 3M	45	0.004
Ag/AgCl Ref 3M	90	0.004
Knitted Noble	0	0.808
Knitted Noble	45	1.081
Knitted Noble	90	0.961
Textile statex	0	0.077
Textile statex	45	0.082
Textile statex	90	0.138
SolGel Dry	0	0.749
SolGel Dry	45	0.644
SolGel Dry	90	0.561
Silicon	0	0.074
Silicon	45	0.030
Silicon	90	0.210
Reusable Spes Medica	0	0.002
Reusable Spes Medica	45	0.011
Reusable Spes Medica	90	0.004

Using bioimpedance method, with injection of current, we notice a significative variation coefficient between the various type of electrodes with p-value = 0.007 but in contrary of intrinsic resistance, the effect of the angle is non-significantly with p=0.884 on the variation coefficient (Table 4).

The injection of current could smoothen the effect of the angle on the anisotropy of materials.

Table. 5 Bipolar characteristic Z (ohm/mm²) obtained for each angle according to each type of electrodes

Electrode type	**Angles °**	**Bipolar Characteristic (ohm/mm²)**
Ag/AgCl Ref 3M	0	8.816
Ag/AgCl Ref 3M	45	6.508
Ag/AgCl Ref 3M	90	8.891
Knitted Noble	0	0.0067
Knitted Noble	45	0.0072
Knitted Noble	90	0.0003
SolGel Dry	0	0.021
SolGel Dry	45	0.025
SolGel Dry	90	0.037
Silicon	0	0.071
Silicon	45	0.103
Silicon	90	0.069
Reusable Spes Medica	0	18.462
Reusable Spes Medica	45	8.813
Reusable Spes Medica	90	13.149
Textile statex	0	0.017
Textile statex	45	0.015
Textile statex	90	0.018

We notice that the angle has no significative effect on the contact impedance (p=0.994) but the type of electrodes stands out as significative difference with p=0.006 (Table 5). We expect that result comparing dry and gel electrode but in fact, the contact impedance is so weak that difference doesn't affect the quadripolar measurement.

As there isn’t difference according to the angle, we can average the values obtained at 0, 45 and 90° to obtain the bipolar characteristic Z (ohm/mm²).

On the other hand, the ranking of the electrodes from least anisotropic to most isotropic differs with and without electric current.

The acquisition frequency does not have a significant effect on the values of X, R and Z, with respectively $p=0.487$, $p=0.999$ and $p=0.999$. Thereafter, the measurements will be made at low frequency (4kHz) to analyse the contact.

On Agar-Agar phantom, with notice a significantly variation coefficient between the bipolar characteristics of the various electrodes ($p<0.001$).

Table.6 Bipolar impedance ranking obtained with each type of electrode

Electrode types	**Z (Ohm/mm3)**
Textile Statex	0.0006
Reusable Spes Medica	0.0008
Silicon	0.0008
SolGel Dry	0.0014
Ag/AgCl Ref3M	0.0023

The difference between the bipolar impedance obtained on phantom with each type of electrode is not significative with p-value<0.0001 (Table 6).

Unexpectedly, bipolar characteristics of gel electrodes (Ag/AgCl Ref 3M) are superior to those of dry electrodes. It could be due to the interaction between both gel (a non-mixed inducing an interface) contrary to the absorption of a part of the Agar-Agar surface with the other dry electrodes or the gel of reusable SpesMedica.

3.1.3. Quadripolar Characteristic on AA Phantom (ohm/mm^3)

We notice a significantly variability coefficient on quadripolar measurement ($p<0.0001$ for each variable, R, X and Z) between the different type of electrodes but the absolute value is relative with the mean of 1.1+/-2.2 % (Figure 6).

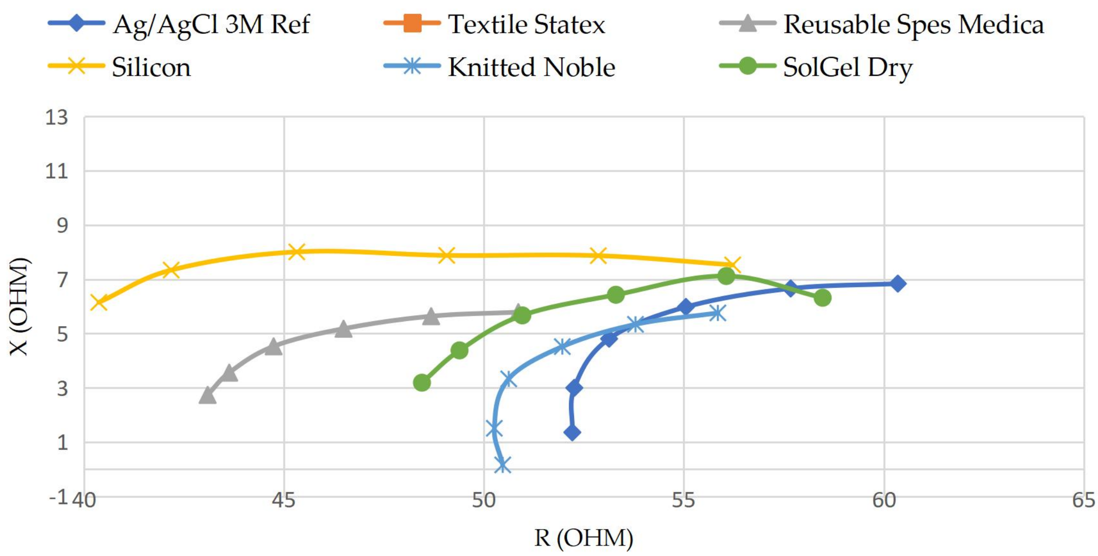


Fig. **6** Bipolar impedance (Cole-cole curve) obtained with each type of electrode

Table. 7 Quadripolar electrical characteristic R, X and Z (ohm) on phantom and those value standardized to the volume analyzed

Fréquence kHz	**R (Ohm)**	**X (Ohm)**	**Z (Ohm)**	**R (Ohm/mm3)**	**X (Ohm/mm3)**	**Z (ohm/mm3)**
4	60.35	6.84	60.74	0.00023	0.00003	0.0002
8	57.67	6.66	58.05	0.00022	0.00003	0.0002
18	55.05	5.98	55.37	0.00021	0.00002	0.0002
40	53.13	4.81	53.35	0.00021	0.00002	0.0002
80	52.26	3	52.35	0.00020	0.00001	0.0002
128	52.21	1.36	52.23	0.00020	0.00001	0.0002
4	55.85	5.76	56.15	0.00062	0.00006	0.0006
8	53.79	5.34	54.05	0.00059	0.00006	0.0006
18	51.96	4.52	52.16	0.00057	0.00005	0.0006
40	50.62	3.33	50.73	0.00056	0.00004	0.0006
80	50.26	1.51	50.28	0.00055	0.00002	0.0006
128	50.47	0.16	50.47	0.00056	0.00000	0.0006
4	58.47	6.32	58.81	0.00048	0.00005	0.0005
8	56.06	7.12	56.51	0.00046	0.00006	0.0005

18	53.3	6.43	53.69	0.00044	0.00005	0.0004
40	50.96	5.67	51.27	0.00042	0.00005	0.0004
80	49.39	4.38	49.58	0.00041	0.00004	0.0004
128	48.45	3.19	48.55	0.00040	0.00003	0.0004
4	56.22	7.53	56.72	0.00022	0.00003	0.0002
8	52.86	7.87	53.44	0.00021	0.00003	0.0002
18	49.07	7.88	49.70	0.00019	0.00003	0.0002
40	45.32	8.01	46.02	0.00018	0.00003	0.0002
80	42.18	7.34	42.81	0.00016	0.00003	0.0002
128	40.37	6.15	40.84	0.00016	0.00002	0.0002
4	50.86	5.79	51.19	0.00020	0.00002	0.0002
8	48.68	5.64	49.01	0.00019	0.00002	0.0002
18	46.49	5.18	46.78	0.00018	0.00002	0.0002
40	44.74	4.53	44.97	0.00017	0.00002	0.0002
80	43.63	3.56	43.77	0.00017	0.00001	0.0002
128	43.09	2.74	43.18	0.00017	0.00001	0.0002
4	59.8	13	61.20	0.00023	0.00005	0.0002
8	54.74	11.96	56.03	0.00021	0.00005	0.0002
18	50	10.27	51.04	0.00019	0.00004	0.0002
40	45.97	8.76	46.80	0.00018	0.00003	0.0002
80	43.6	6.9	44.14	0.00017	0.00003	0.0002
128	42.42	5.38	42.76	0.00016	0.00002	0.0002

3.2. Characterization on human quadriceps

We notice on the repeatability of Zcontact a significantly difference between the various electrodes but with a relative variability coefficient of 1.8+/-2.8 % (Table 8).

Table. 8 Comparison of Z-Contact (ohm.mm3) for each gender

Woman	**Zcontact (ohm/mm3)**	**Man**	**Zcontact (ohm/mm3)**

Silicon	0.001428049	Reusable SpesMedica	0.001305614
Reusable SpesMedica	0.001715072	Silicon	0.002042707
Textile Statex	0.003540037	Textile Statex	0.003158221
Ag/AgCl Ref 3M	0.00511322	Ag/AgCl Ref 3M	0.00400333
SolGel Dry	0.008217425	SolGel Dry	0.006400453
Knitted Noble	0.014826908	Knitted Noble	0.017209603

As expected, the dry electrodes have a higher impedance contact than the gel electrode except Statex's textile. The ranking is similar between the two gender even if the skin is different.

3.3.2. Quadripolar characteristics

We notice on the repeatability of impedance a significantly difference between the various electrode with p-value<0.0001 but with a relative mean Cv of 0.03+/-2.4 %.

Except Reusable Spesmedica, wich is another gel electrode (p-value = 0.41), all the others electrodes obtained significantly difference of electrical characteristic with the reference Ag/AgCl 3M (Figure 7).

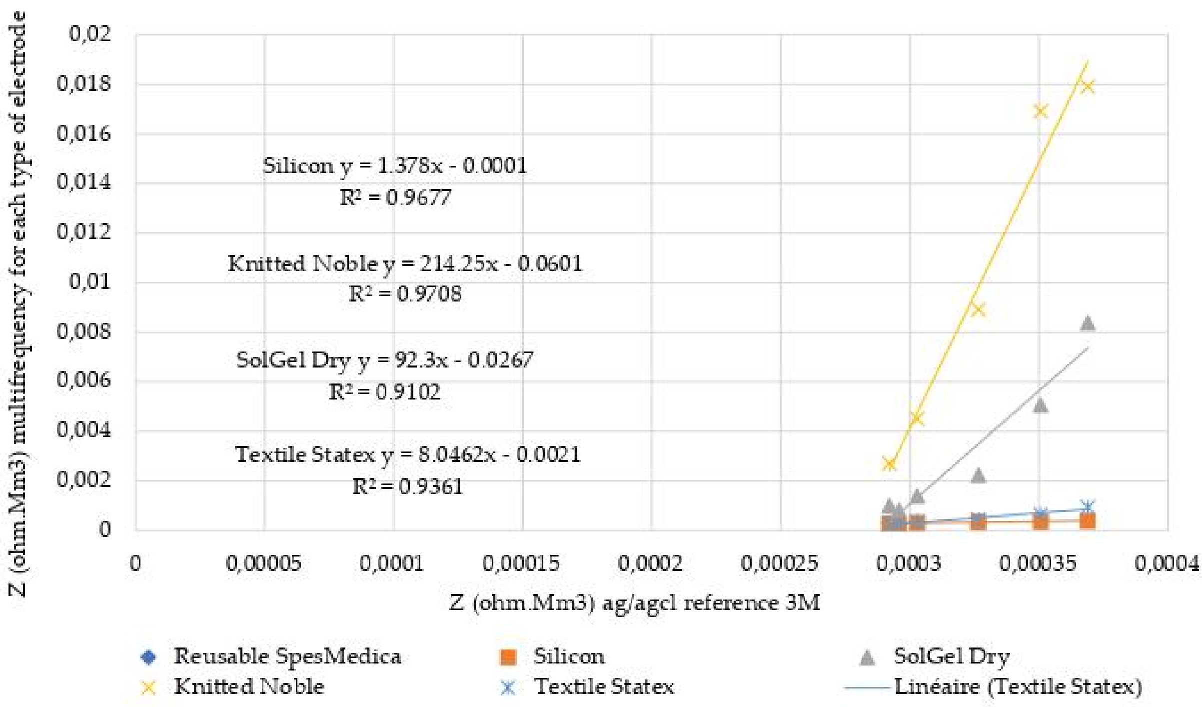


Fig. 7 Comparison on Z (ohm.mm3) obtained on woman quadriceps according to each type of electrodes versus Ag/AgCl Reference

On woman, we notice an R^2 dispersion up than 0.90 for each type of electrodes suggesting that it's possible to correct each type of electrodes even if the accuracy is significantly different than the 3M ref. On man, excepted Textile Statex, we obverse the same.

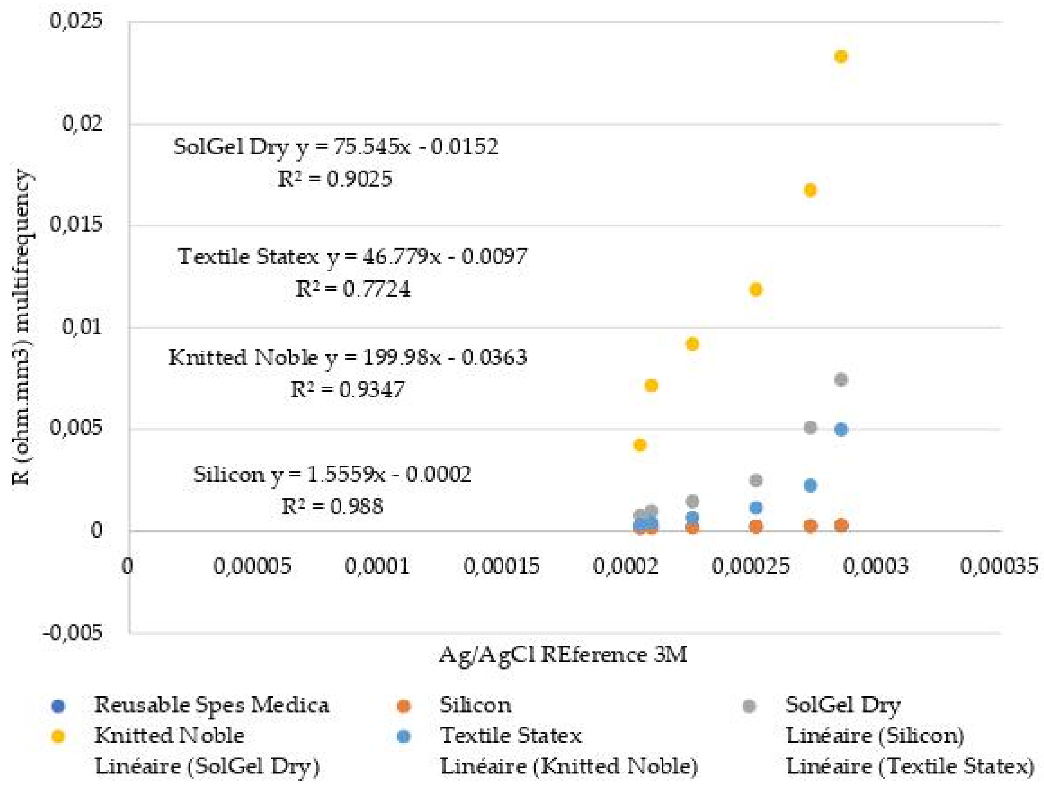


Fig. 8 Comparison on Z (ohm.mm3) obtained on man quadriceps according to each type of electrodes versus Ag/AgCl Reference

The Equations (3)–(2) are used to determine the corresponding corrected value using Dry electrodes as Ag/AgCl gel electrodes, according to gender.

For Zcorr (ohm)

For Woman

Textile Statex $y = 8.0462x - 0.0021$ ($R^2 = 0.9361$) (3)

SolGel Dry $y = 0.0101x + 0.0003$ ($R^2 = 0.9227$) (4)

Knitted Noble $y = 0.0045x + 0.0003$ ($R^2 = 0.9708$) (5)

Silicon $y = 0.7022x + 9E\text{-}05$ ($R^2 = 0.9677$) (6)

Reusable SpesMedica $y = 0.9415x$ ($R^2 = .9995$) (7)

For Man

Textile Statex $y = 3E\text{-}05\ln(x) + 0.0005$ ($R^2 = 0.9729$) (8)

SolGel Dry $y = 4E\text{-}05\ln(x) + 0.0005$ ($R^2 = 0.9937$) (9)

Knitted Noble $y = 0.0047x + 0.0002$ ($R^2 = 0.9347$) (10)

Silicon $y = 0.635x + 1E\text{-}04$ ($R^2 = 0.988$) (11)

Reusable SpesMedica $y = 1.0091x + 7E\text{-}06$ ($R^2 = 0.9932$) (12)

For Rcorr (ohm)

For Woman

Textile Statex $y = 45.505x – 0.0011$ ($R^2 = 0.9264$) (13)

SolGel Dry $y = 3E\text{-}05\ln(x) + 0.0005$ ($R^2 = 0.9781$) (14)

Knitted Noble $y = 0.0041x + 0.0003$ ($R^2 = 0.9577$) (15)

Silicon $y = 0.6932x + 9E\text{-}05$ ($R^2 = 0.9638$) (16)

Reusable SpesMedica $y = 0.7806x + 6E\text{-}05$ ($R^2 = 0.9626$) (17)

For Man

Textile Statex $y = 3E\text{-}05\ln(x) + 0.0005$ ($R^2 = 0.9382$) (18)

SolGel Dry $y = 3E\text{-}05\ln(x) + 0.0004$ ($R^2 = 0.9856$) (19)

Knitted Noble $y = 0.0043x + 0.0002$ ($R^2 = 0.8882$) (20)

Silicon $y = 0.7x + 9E\text{-}05$ ($R^2 = 0.9918$) (21)

Rezsable SpesMedica $y = 0.9914x + 1E\text{-}05$ ($R^2 = 0.9915$) (22)

For Xcorr (ohm)

For Woman

Textile Statex $y = 5.1353x + 0.0002$ ($R^2 = 0.059$) (23)

SolGel Dry $y = (2E\text{-}05\ln(x) + 0.0002)/2$ ($R^2 = 0.605$) (24)

Knitted Noble $y = 0.002x + 3E\text{-}05$ ($R^2 = 0.3274$) (25)

Silicon $y = 2.0998x - 7E\text{-}05$ ($R^2 = 0.9704$) (26)

Reusable SpesMedica $y = 1.8909x - 5E\text{-}05$ ($R^2 = 0.5523$) (27)

For Man

Textile Statex y = -12.967x2 + 0.0359x + 2E-05 (R^2 = 0.5487) (28)

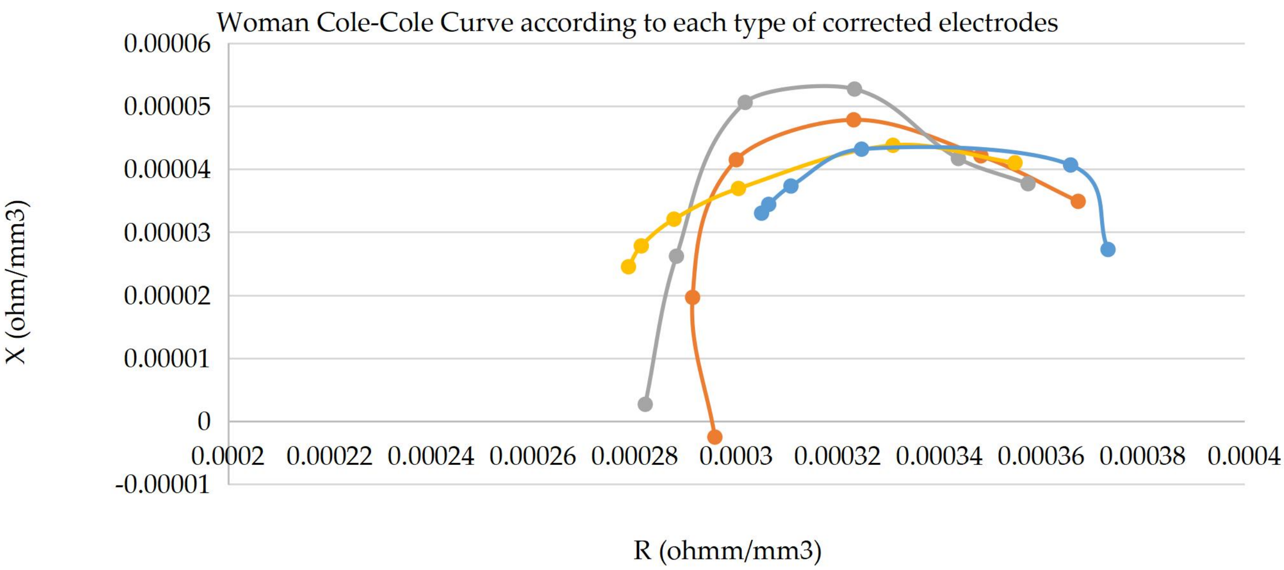
SolGel Dry y = 8E-06ln(x) + 8E-05 (R^2 = 0.1974) (29)

Knitted Noble y = 0.0032x + 9E-06 (R^2 = 0.4652) (30)

Silicon y = -22438x2 + 2.9483x - 5E-05 (R^2 = 0.9492) (31)

Reusable SpesMedica y = 1.0517x - 3E-06 (R^2 = 0.6897) (32)

As we obtain very different range of value for Textile statex (about 10-3 ohm/mm3 instead of 10-5 ohm/mm3) we can't graph it with the other electrodes. Furthermore, we didn't obtain a curve but a line for this type of electrodes.



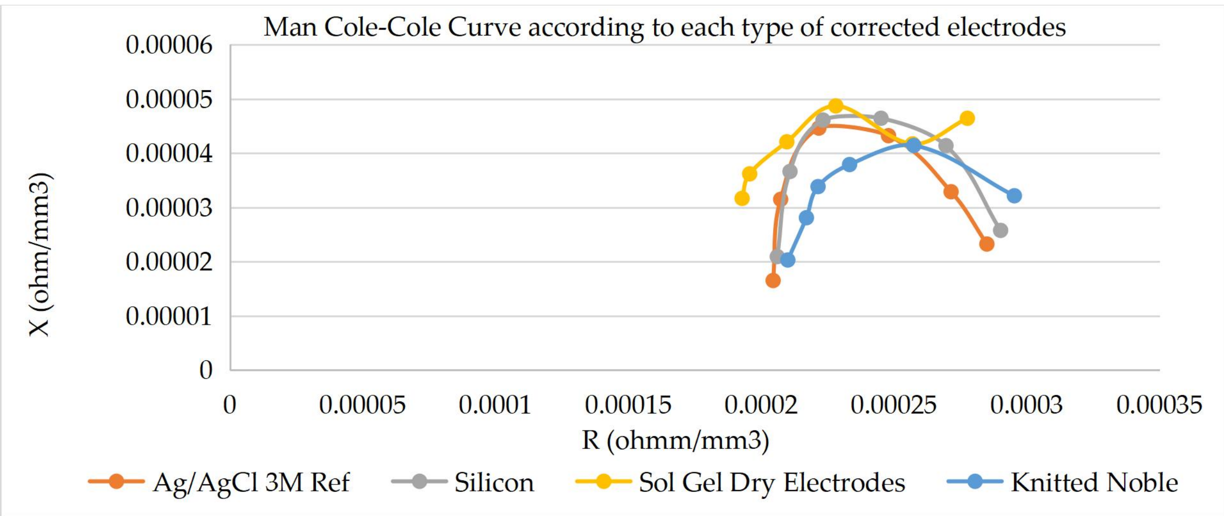


Fig. 9 Cole-Cole curves obtained for each type of electrodes respectively on woman and man quadriceps

We notice none significantly difference between Rcorr (Ohm/mm3) and Xcorr (Ohm/mm3), respectively p-value = 0.952 and 0.112 used to graph the Cole-Cole curve.

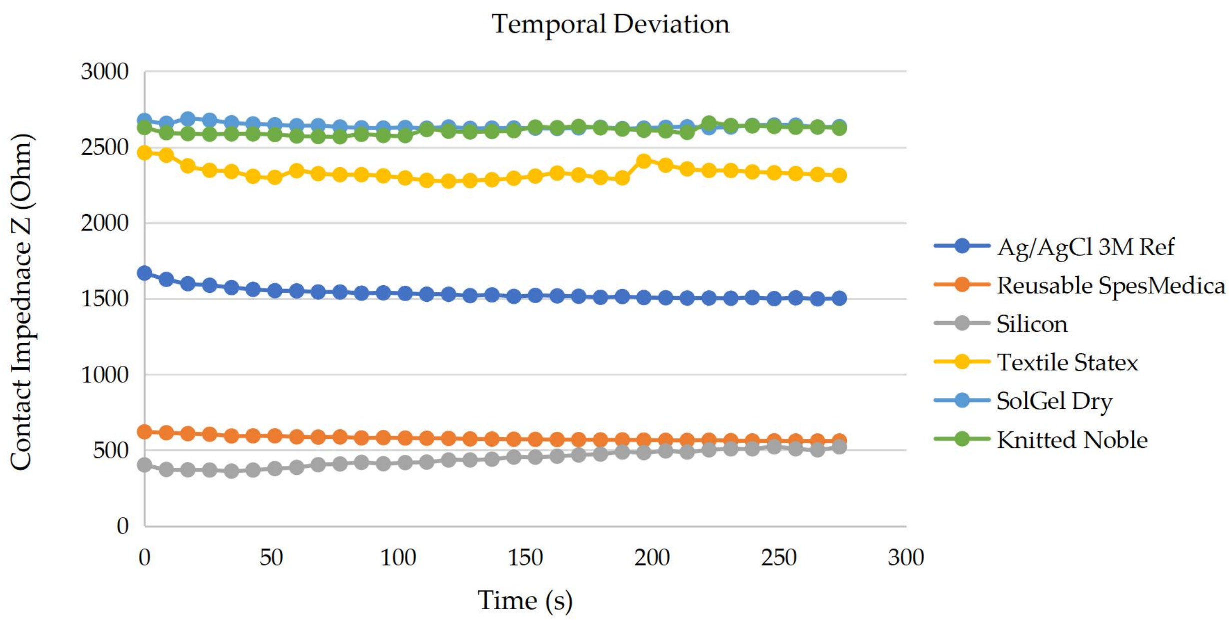


Fig. 10 Temporal deviation of contact impedance Z (ohm) on human skin according to each type of electrodes

4. Discussion

The electrodes investigated in this study differ in their contact surface area, which introduces a potential source of bias in the comparison of their electrical properties.

4.1. Difference between AA phantom and human measurements

The discrepancies observed between the results obtained on the AA phantom and those measured on human subjects can be explained by several factors.

The AA phantom is designed to reproduce the electrical properties of human quadriceps tissue and exhibits anisotropic behavior at 0° and 90°, consistent with values reported in the literature. However, it does not fully replicate the complexity of biological tissues. In particular, the skin layer covering the muscle is not represented, despite the fact that different tissues exhibit distinct electrical properties [29]. Consequently, the electrode–phantom interface differs substantially from the electrode–skin interface.

Furthermore, temperature and moisture conditions differ between the two experimental setups. Human skin is typically at approximately 30°C, whereas the phantom is measured at room temperature. In addition, variations in skin hydration and perspiration are not reproduced in the phantom model [28]. These differences in thermal and moisture conditions may contribute to the discrepancies observed in the results.

Another important factor is the pressure applied at the electrode interface. For phantom measurements, a controlled pressure of 0.26 g/cm² was applied, as defined in a previous study. In contrast, during human experiments, the applied pressure was not quantitatively controlled but subjectively adjusted by the participants to mimic the sensation of a tight-fitting garment (e.g., leggings). This difference in contact conditions may also influence the measurements.

4.2. Need for a larger human population

To further validate these findings, the study should be extended to a larger and more diverse population. Future investigations should include participants with varying characteristics, such as sex, training status (trained vs. untrained), body morphology, and ethnicity. This would allow assessment of the robustness and generalizability of the selected electrode and determine whether specific electrode types are more suitable for particular populations.

5. Conclusion and perspectives

In this study, we characterized a panel of dry electrodes, first on an AA phantom and subsequently on two human subjects, one male and one female. This characterization provides an initial framework for selecting electrodes based on their electrical properties.

As a second step, we proposed correction equations that allow textile electrodes to be referenced against conventional gel electrodes, commonly used as a standard. This approach enables an initial selection of electrodes and their use in a manner comparable to gel electrodes.

For applications in connected garments, further investigations are required regarding textile processability, durability to washing, and adherence to the skin. Taking these factors into account, silicone electrodes are recommended for wearable applications.

The main contribution of this study lies in demonstrating that electrodes can be selected according to various criteria while maintaining compatibility with gel-electrode-based measurements. This provides flexibility in the design of smart garments while ensuring reliable bioimpedance monitoring.

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