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A new method to assess skin treatments for lowering the impedance and noise of individual gelled Ag–AgCl electrodes

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Abstract

A model-based new procedure for measuring the single electrode–gel–skin impedance (Z_{EGS}) is presented. The method is suitable for monitoring the contact impedance of the electrodes of a large array with limited modifications of the hardware and without removing or disconnecting the array from the amplifier. The procedure is based on multiple measurements between electrode pairs and is particularly suitable for electrode arrays. It has been applied to study the effectiveness of three skin treatments, with respect to no treatment, for reducing the electrode–gel–skin impedance (Z_{EGS}) and noise: (i) rubbing with alcohol; (ii) rubbing with abrasive conductive paste; (iii) stripping with adhesive tape. The complex impedances Z_{EGS} of the individual electrodes were measured by applying this procedure to disposable commercial Ag–AgCl gelled electrode arrays (4×1) with a $5\,mm^2$ contact area. The impedance unbalance $\Delta Z = Z_{EGS1} - Z_{EGS2}$ and the RMS noise (V_{RMS}) were measured between pairs of electrodes. The tissue impedance Z_T was also obtained, as a collateral result. Measurements were repeated at $t_0 = 0\,min$ and at $t_{30} = 30\,min$ from the electrode application. Mixed linear models and linear regression analysis applied to Z_{EGS}, ΔZ and noise V_{RMS} for the skin treatment factor demonstrated (a) that skin rubbing with abrasive conductive paste is more effective in lowering Z_{EGS}, ΔZ and V_{RMS} ($p < 0.01$) than the other treatments or no treatment, and (b) a statistically significant decrement ($p < 0.01$), between t_0 and t_{30}, of magnitude and phase of Z_{EGS}.

Rubbing with abrasive conductive paste significantly decreased the noise V_{RMS} with respect to other treatments or no treatment.

Keywords: EMG, electrode impedance, electrode noise, skin treatments

(Some figures may appear in colour only in the online journal)

1. Introduction

Detecting electromyographic (EMG) signals from the skin of human subjects, by means of electrode arrays, is becoming a commonly used method in neuromuscular research. In particular, high density surface electromyography (HDsEMG) is applied for the identification of anatomical and physiological muscle features (innervation zone, muscle fiber length and orientation, muscle fiber conduction velocity), and the decomposition of the EMG signal into the constituent motor unit action potential trains (Holobar *et al* 2009, Merletti *et al* 2010a, 2010b). The clinical acquisition of the HDsEMG signals with high signal-to-noise and signal-to-interference ratios is therefore of prime importance but contrasted by the high impedance and noise due to small electrode contact size. To detect high quality signals in HDsEMG applications, the electrodes must: (a) be small enough (5–10 mm^2) to fit in arrays or matrices with small inter-electrode distance, typically 5–10 mm; (b) have low and similar impedances; (c) have low noise levels.

The impedances of any two single electrodes used for the acquisition of single differential (SD) signals must have similar values in order to minimize the differential power line voltage that can be injected in the acquisition system by the 'voltage-divider effect' described in figure 1(*a*).

The fraction of common mode voltage (V_{cm}), due to parasitic coupling with the power line, converted into differential voltage V_d, is proportional to the impedance unbalance $\Delta Z = Z_{EGS1} - Z_{EGS2}$ (Degen 2007, Merletti *et al* 2009, 2010a). Consider an amplifier with input capacitance of 10 pF in parallel to a resistance >1 GΩ. Its input impedance at 50–60 Hz would be about 300 MΩ. A common mode voltage V_{cm} = 1 V_{RMS} and ΔZ = 0.1 MΩ would produce a differential signal V_d = 0.33 mV_{RMS}, which is comparable with the EMG signal magnitude (from a few μV to a few mV).

For this reason, lowering and matching the single electrode impedances, as well as testing their stability in time, is a key factor for (a) increasing the signal-to-interference ratio (Grimnes 1983, Degen *et al* 2007), and (b) limiting the number of 'bad channels' in a high density EMG electrode array (Merletti *et al* 2010a). Previous studies have been conducted mainly on large electrodes, for ECG application, (Grimnes 1983, Fernández and Pallás-Areny 2000, Searle *et al* 2000, Hewson *et al* 2003, Puurtinen *et al* 2006, Riistama and Lekkala 2006, Degen *et al* 2007, Gruetzmann *et al* 2007, Tronstad *et al* 2010, Meziane *et al* 2013), not suitable for HDsEMG applications. Some studies were focused on the measurement of only the ΔZ (Spinelli *et al* 2006, Degen *et al* 2007), and did not investigate the effect of different skin treatments.

It is known that the impedance of a pair of gelled electrodes decreases with time (Searle *et al* 2000, Hewson *et al* 2003, Grimnes *et al* 2008), but the behavior of single impedance magnitude and phase of the impedance unbalance ΔZ were not investigated for small electrodes. In particular, the method proposed by Grimnes (Grimnes 1983) measures the individual electrode impedance but the required hardware cannot be easily adapted to the sequential scanning of a large electrode array.

The purpose of this work is: (a) to present a method that is suitable for measuring the impedance of individual electrodes of an array in a clinically viable situation, without disconnecting the array from the amplifiers, and (b) to use it to compare the effectiveness of three clinically popular skin treatments for lowering Z_{EGS}, ΔZ (at 50 Hz) and noise, with respect to no treatment (NT). The three skin treatments are: (a) rubbing with alcohol (RA); (b) rubbing with abrasive conductive paste (Everi, by Spes Medica) (RP); (c) stripping (five times) with new pieces of adhesive tape (ST) (staroffice, code: 909027).

Preliminary results of this work have been presented at the ICNR 2012 (Merletti *et al* 2012).

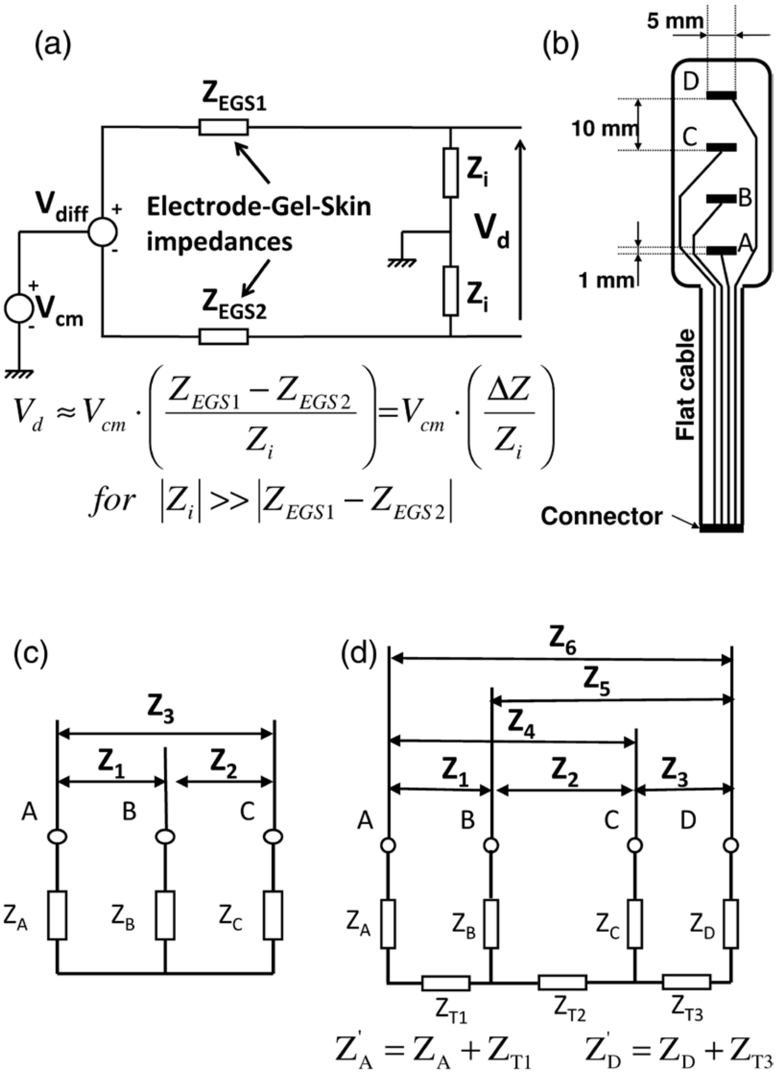

Figure 1. (*a*) The difference between Z_{EGS1} and Z_{EGS2} unbalances the voltage dividers $Z_i/(Z_{EGS1} + Z_i)$ and $Z_i/(Z_{EGS2} + Z_i)$, where Z_i is the input impedance of the amplifier, converting a fraction of V_{cm} (power line interference) into V_d. (*b*) Adhesive electrode array in Ag/AgCl (Spes Medica, Italy) with inter-electrode distance IED = 10 mm, single electrode area is 5 mm^2. The coupling between the electrodes and the skin is provided by a perforated double adhesive foam (1 mm thick) with cavities filled with conductive gel (1% KCl, 3% HEC, 1% propylene glycol, 95% preserved water). (*c*) and (*d*) Schematic representation of the measuring system with the 3-electrode model and with the 4-electrode model. Z_A, Z_B, Z_C and Z_D are the four EGS impedances, Z_{T1}, Z_{T2} and Z_{T3} are the impedances of the tissue. Only Z_{T2} can be estimated. Z_1, Z_2, Z_3, Z_4, Z_5 and Z_6 are the impedances between the six pairs of electrodes, sequentially measured by means of the impedance meter depicted in figure 2(a).

2. Materials and methods

2.1. A new method to estimate individual electrode impedances

Figure 1(*b*) depicts the commercially available disposable 4-contact array for EMG detection used in this work. The array was developed at LISiN (and manufactured by Spes Medica, Battipaglia, Italy) and is made of four Ag–AgCl gelled electrodes (4 × 1 array, electrode surface of 5 mm^2, inter-electrode distance (IED) = 10 mm). It was applied to the skin by means of a double adhesive foam, 1 mm thick, with 5 mm^2 rectangular holes filled with gel and aligned with the Ag–AgCl contacts.

Homemade conductive gel (composition: 1% KCl, 3% HEC, 1% propylene glycol, 95% preserved water) was used for all the measurements. This is an easy to make common type of gel, and easy to make. In view of future comparisons with other gels, it was preferred to commercially available gels because their composition is usually not known.

Consider the very simple 3-contact model depicted in figure 1(*c*) where A, B, C, represent the electrodes and Z_A, Z_B, Z_C, represent their contact impedances connected together by the sub-cutaneous layers, whose transversal impedance is assumed to be negligible with respect to $Z_{A,B,C}$ and is not considered in this model. The three measured inter-electrode impedances, Z_1, Z_2, Z_3, are related to Z_A, Z_B, Z_C by the linear system of equation (1) which can be easily inverted to obtain the individual contact impedances.

$$\begin{cases} Z_1 = Z_A + Z_B \\ Z_2 = Z_B + Z_C \\ Z_3 = Z_A + Z_C \end{cases} \tag{1}$$

If four electrodes (A, B, C, D) are used, the model described by equation (1) is expanded into a new system of six equations in four unknowns. The associated 6 × 4 matrix has rank 4 and the system can be solved either by pseudo-inversion or by performing only four independent measurements.

Figure 1(*d*) shows a better (but still approximated) model accounting for the transversal discrete tissue impedances Z_{T1}, Z_{T2} and Z_{T3}, where Z_{T1} and Z_{T3} cannot be separated from Z_A and Z_D. In this case, a system of six equations in five unknowns is obtained. The model can be written in matrix form (equation (2)) as $\boldsymbol{m} = \mathbf{A}\boldsymbol{x}$, where $\boldsymbol{m}$ is the measurement vector and $\boldsymbol{x}$ is the unknown vector of complex impedances:

$$\underbrace{\begin{bmatrix} Z_1 \\ Z_2 \\ Z_3 \\ Z_4 \\ Z_5 \\ Z_6 \end{bmatrix}}_{\underline{\mathbf{m}}} = \underbrace{\begin{bmatrix} 1 & 1 & 0 & 0 & 0 \\ 0 & 1 & 1 & 0 & 1 \\ 0 & 0 & 1 & 1 & 0 \\ 1 & 0 & 1 & 0 & 1 \\ 0 & 1 & 0 & 1 & 1 \\ 1 & 0 & 0 & 1 & 1 \end{bmatrix}}_{\mathbf{A}} \underbrace{\begin{bmatrix} Z'_A \\ Z_B \\ Z_C \\ Z'_D \\ Z_{T2} \end{bmatrix}}_{\underline{\mathbf{x}}} \tag{2}$$

where $Z_A{}' = Z_A + Z_{T1}$ and $Z_D{}' = Z_D + Z_{T3}$.

Matrix A has rank 5 because one equation is a linear combination of three other equations (e.g. $Z_2 = Z_4 + Z_5 - Z_6$) and the system can be solved by matrix pseudo-inversion or by inversion after removing the dependent equation. However, since noise is present in the measurements, equation (2) may not be exactly satisfied, therefore a solution in the 'least square' sense is preferred, by computing the pseudo-inverse of A,

$$A^{\dagger} = (A^T A)^{-1} A^T \quad (3)$$

$$x = A^{\dagger} m \quad (4)$$

This approach provides only the value of Z_{T2}. However, given a longer or a 2D electrode array, the process may be iterated by shifting the electrode quadruplet (A, B, C, D) to obtain Z_{T1} and Z_{T3}, as well as Z_A and Z_D. Since the physical system is continuous and not discrete, Z_T is a poorly defined model component or quantity. However, its separation is expected to provide a more accurate approximation of the electrode–gel–skin impedance than is given by equation (1), and therefore a better estimate of ΔZ.

2.2. Impedance measurements

Ten healthy subjects participated in the experiments. The local ethics committee approved all experimental recordings and all the subjects signed an informed consent form before participation.

Impedance measurements were conducted by using an impedance meter, designed at LISiN and described in figure 2(*a*). The instrument injects current through the impedance and measures the resulting voltage. A voltage input signal V_I, provided by a signal generator (Agilent 33210A), is converted into a proportional current signal I_I (proportionality factor $k = 10^{-6}\,\mathrm{A\,V^{-1}}$). Both V_I and the voltage drop across the impedance V_V are simultaneously acquired using the National Instruments NI-DAQ USB-6210 acquisition board (A/D conversion on 16 bits, input voltage range −10–10V, $f_{\mathrm{Sampling}} = 10\,\mathrm{kHz}$). A sinusoidal current $I_I = 200\,\mathrm{nA_{PP}}$ with a 10s frequency sweep between 2 Hz and 1000 Hz is applied between an electrode pair. The value of the impedance at each frequency is obtained from the ratio of the Fourier transforms FFT(V_V) / FFT(V_I).

The individual electrode complex impedance Z_{EGS} was measured, using the algorithm described by equations (2)–(4), for the six contact pairs of figure 2(*d*), and the impedance unbalance ΔZ between such contacts were measured at $t_0 = 0\,\mathrm{min}$ and at $t_{30} = 30\,\mathrm{min}$ after electrode application. The relative contribution of the tissue impedance Z_T to the impedance between two electrodes was examined.

The spectral lines in a neighborhood of ±10 Hz of all the 50 Hz harmonics were removed in order to avoid power line interference. 24 impedance values in the selected range 2 Hz–1000 Hz were calculated by selecting 24 frequency intervals and computing the mean of the impedance values in each interval. The magnitude and phase of the impedance at 50 Hz were obtained by interpolation between the values measured at lower and higher frequencies.

The real part of each impedance between a pair of electrodes, was correlated to the noise generated by the same electrode pair to verify if electrode noise is due to thermal noise (Riistama and Lekkala 2006).

2.3. Validation of the impedance estimation algorithm

The model was tested by applying the system of figure 2 and the algorithm of equations (2)–(4) to known R-C groups arranged as Z_A', Z_B, Z_C, Z_D' and Z_{T2} (figure 1(*d*)). The resistors range was 0.1–1 MΩ for the Z_{EGS} and 10–20 kΩ for the Z_T, with tolerances of 1%. The capacitors range was 1–10 nF, with tolerances of 10%. The real and the imaginary part of the impedances were estimated with errors lower than 8.6% (mean value: 1.37%), within the range of the component tolerance. This procedure validates the algorithm, not necessarily the discrete model of what is actually a continuous system.

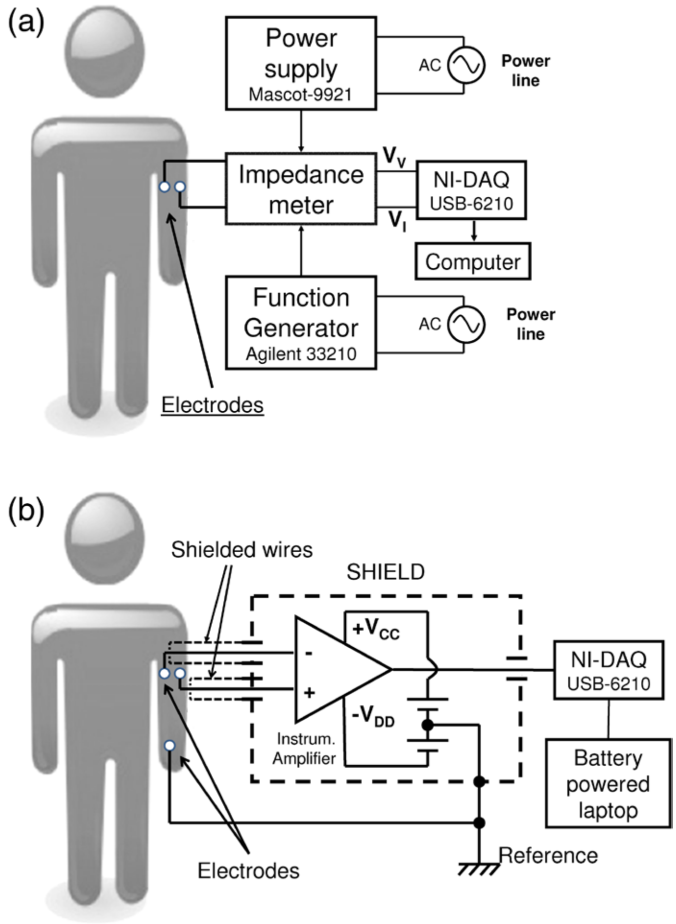

Figure 2. (*a*) Experimental acquisition setup for the impedance measurements. The power supply (Mascot 9921) is a medical grade device. (*b*) Schematic of the experimental measurement setup for the noise measurements. The noise signal was amplified by means of a low noise amplifier and it was sampled with NI-DAQ USB-6210 (National Instrument Data Acquisition System). The wires attached to the electrodes and the amplifier were shielded. The EGS noise is calculated, assuming that V_{EGS} and V_{instr} are uncorrelated, as $V_{\mathrm{EGS-RMS}} = \sqrt{V_{\mathrm{tot}}^2 - V_{\mathrm{RMS\_instr}}^2}$ where V_{tot} is the total measured noise (RMS) and $V_{\mathrm{RMS\text{-}instr}}$ is the instrumentation noise (voltage and current noise) measured on a 500 kΩ resistor.

2.4. Noise measurements

The electrode array depicted in figure 1(*b*) was applied on a totally relaxed biceps muscle and electrode noise was measured using a low noise amplifier designed at LISiN and described in figure 2(*b*) (gain = 10 568, bandwidth: 10 Hz–1 kHz), and the NI-DAQ USB-6210 acquisition board. For each acquisition, 10 s of noise were recorded and stored (sampling rate = 10 kHz). A single 1 s epoch was selected by visual inspection as that showing absence of spikes (such as due to EMG) or low frequency fluctuation around the zero baseline. The power line interference in the selected epoch was removed offline by means of the 'spectral interpolation' technique. With this technique, N spectral lines in a neighborhood (±2 Hz) of the 50 Hz harmonics in the Fourier transform of the signal (magnitude and phase) are replaced with the value obtained by linear interpolation between the edges of the deleted neighborhood. In this case $N = 5$. The inverse Fourier transform is then computed to recover a 'clean' signal. Output noise values were divided by the gain G of the noise amplifier and digitally filtered by means of a second order Butterworth band-pass filter with cutoff frequencies 10 Hz and 1000 Hz. Finally, the RMS value of the EGS interface noise was computed by quadratic subtraction of the RMS noise value of the instrumentation (measured value $V_{\mathrm{RMS\_instr}} = 0.86\ \mu V_{\mathrm{RMS}}$) from the RMS value of the measurement (figure 2(*b*)). The RMS noise value of the instrumentation ($V_{\mathrm{RMS\_instr}}$) was measured by placing a 500 kΩ resistor between the input pins of the noise amplifier and acquiring the output signal by means of the NI-DAQ USB-6210 acquisition board. This measurement accounts for both the voltage and the current noise of the amplifier due to a resistor whose nominal value is comparable with the impedance magnitude of the EGS interface measured at 50 Hz. The process presumes that the instrumentation and electrode noises are uncorrelated. The epoch duration for the calculation of the RMS values was 1 s.

2.5. Experimental setup

In order to reduce experimental time, the four treatments (NT, RA, RP, ST) were applied to four portions of the same biceps of each subject and one electrode array was placed on each portion. To verify that the portions of the skin had the same properties, the following procedure was implemented on seven test subjects. The length of the biceps brachii (insertion points of tendons, detected by palpation of the ends of the muscle) was washed with soap, rinsed and divided into five equal proximal to distal regions. A 4 × 1 electrode array (figure 1(*b*)) was then placed transversally on each region. The impedance of the single EGS interfaces Z_B and Z_C was measured with the 4-electrode procedure (figure 1(*d*) and equations (2)–(4)) in each region. Statistical analysis (ANOVA and Newman–Keuls *post hoc* test) of the impedance values in the band 30–500 Hz showed significant differences at the lower frequencies ($p < 0.01$ at 30 Hz) only between the impedance values at the most distal region and those of the other regions. Therefore, only the four proximal regions of the biceps were considered for the study.

During the measurement procedure, for each of the ten subjects, each of the four considered regions of the biceps was treated differently (NT, RA, RP, ST), randomizing the combinations of section and skin treatment. Although, for research purposes, adhesive tape stripping has been applied up to 12 or 15 times (Yamamoto 1976, de Talhouet and Webster 1996) the selected tape (staroffice, code: 909027) used for stripping (ST) was applied and removed five times (a new piece of tape every time) to reduce time and skin damage; after this treatment the skin was slightly reddened and examination of the removed tape with a magnifying glass showed almost no speckles.

After the skin treatment of each region, the 4 × 1 electrode array was applied. The impedance measurements, for each treatment, were then carried out on the six electrode pairs (figures 1(*d*) and 2(*a*)) within 3 min ($t_o = 0$ min). After the impedance measurements, noise acquisitions (10 s of noise recordings) from the same six electrode pairs were carried out as in figure 2(*b*).

During the acquisition the subjects were instructed to maintain the arm in a resting position and completely relax the biceps. The impedance and the noise of each electrode pair (and for each treatment) were measured at the application time ($t_0 = 0$ min) and after 30 min ($t_{30} = 30$ min). The central EGS impedance values (Z_B and Z_C in figure 1(*d*)), the tissue impedance Z_{T2} and the impedance unbalance $\Delta Z = Z_B - Z_C$ were estimated from these data, as described in 2.2 and 2.4.

2.6. Statistical analysis

Statistical analysis was applied to the values of:

(a) magnitude and absolute phase of the single EGS impedances Z_B and Z_C (two values of Z for each subject, each time and each skin treatment) at 50 Hz, to test the effect of time and of treatment on Z_{EGS}. Mixed model analysis, using 'R' and a linear regression analysis, considering 'treatment' and 'time' as factors and 'subject' as random or fixed additive effect, were respectively used. The effects of the factor 'position' (B and C in figure 1(*b*)) and all the interactions involving 'position' were non-statistically significative (in both statistical approaches) and the observations Z_B and Z_C were considered as replicate measures of the same quantity. The statistical analyses were therefore carried out neglecting the effect of the 'position' factor and considering the mean of the two statistically non-different impedances, in order to have one observation per condition for each subject.

(b) impedance unbalance $\Delta Z = Z_B - Z_C$ at 50 Hz (one value of ΔZ for each subject, each time and each skin treatment).

(c) tissue impedance Z_T at 50 Hz (one value of Z_T for each subject, each time and each skin treatment).

(d) noise (V_{RMS}), measured in the bandwidth 10–1000 Hz. Ten subjects were tested; subject #1 was excluded from the analysis since there was presence of EMG activity in the noise measurements. Six repeated and independent measurements were performed sequentially for the six electrode pairs of figure 1(*d*), for each time, each skin treatment and each subject. Mixed model analysis and a linear regression analysis were performed considering the mean of the six statistically non-different noise measures (therefore one observation per condition for each subject) as dependent variable and 'treatment', 'time' and 'subject' as factors. 'Subject' was a random factor in the mixed model and a fixed factor in the linear regression model. The effects of the factor 'position' (six electrode pairs in figure 1(*d*)) and all the interactions involving 'position' were verified as non-statistically significative (in both statistical approaches) and the six noise measurements were considered as replicate measures of the same quantity, and averaged.

The data obtained from the ten subjects were analyzed by means of the software 'STATISTICA—Statsoft' and 'R' (Bates *et al* 2014). The normality of the data distribution was tested with the Shapiro–Wilks test. Depending on the result, the ANOVA or the ANOVA by ranks test was implemented (ways: treatment and time, with subjects as additive fixed effect) with significance levels set at 0.05. In order to confirm the statistical analysis, a mixed model analysis with subjects as a random additive effect was also implemented, providing the same results.

To verify if the noise generated by an electrode pair is associated to the Johnson noise of the resistive part of the impedance, the relation between the real part of the impedance of an electrode pair, $\mathrm{Re}\{Z_{pair}\}$ measured at 50 Hz, and the RMS noise values between the same pair was investigated at t_0 and t_{30}. The Wilcoxon signed rank test was performed on the change of V_{RMS} versus $\mathrm{Re}(Z_{pair})$ between t_0 and t_{30} $\left(\frac{\Delta V_{RMS}}{\Delta \mathrm{Re}\left(Z_{pair}\right)} = \frac{V_{RMS}(t_{30}) - V_{RMS}(t_0)}{\mathrm{Re}\left\{Z_{pair}(t_{30})\right\} - \mathrm{Re}\left\{Z_{pair}(t_0)\right\}}\right)$, for all the subjects ($N = 9$) and all the electrode pairs ($N = 6$), to verify if there is a statistically significant common behavior with time for the two quantities. This approach must be considered as qualitative since $\mathrm{Re}(Z_{pair})$ is not a resistor (it is function of frequency).

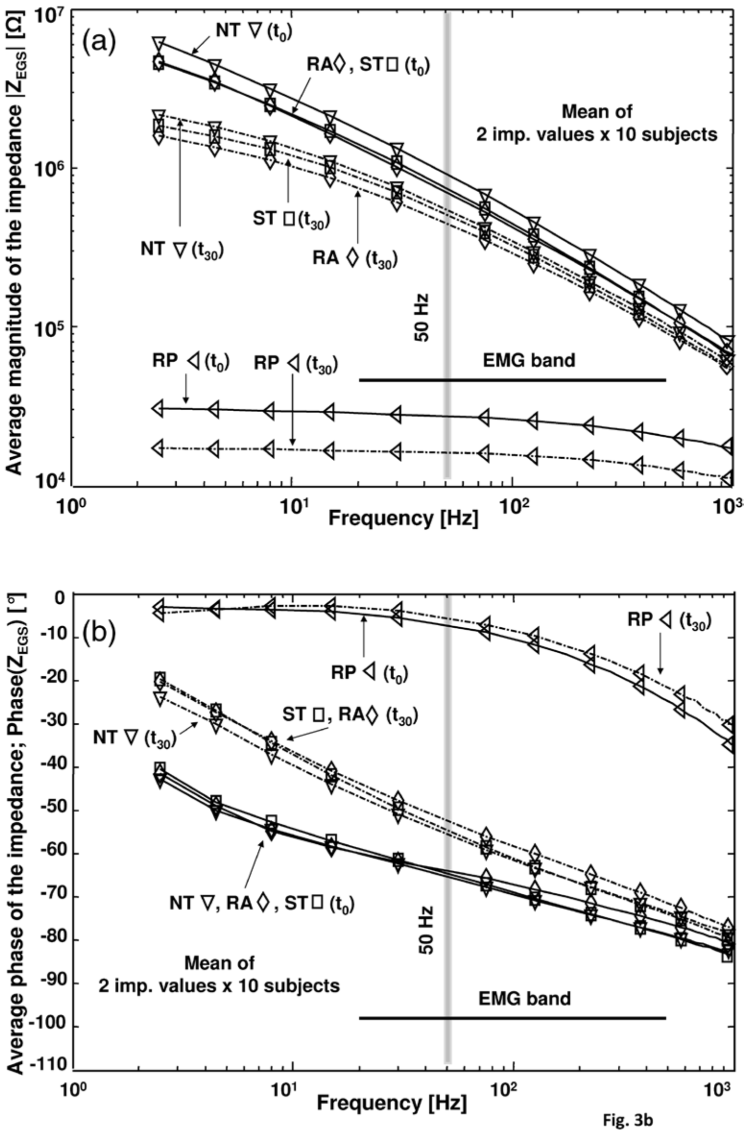

Figure 3. (*a*) Mean values of the EGS impedance magnitudes averaged across electrodes ($N = 2$) and subjects ($N = 10$) for the four different treatments, at t_0 and at t_{30}. Treatments: no treatment (NT), rubbing with alcohol (RA), rubbing with abrasive conductive paste (RP), stripping (five times) with new pieces of adhesive tape (ST) (staroffice, code: 909027). (*b*) Mean values of the EGS impedance phases. See sections 2.2, figures 1 and 2(a) for electrode description and estimation technique.

3. Results

3.1. Impedance measurements

The mean values of Z_{EGS} (magnitude and phase) computed across the electrodes ($N = 2$) and the subjects ($N = 10$) are reported in figure 3 as a function of frequency for t_0 and t_{30} and for the four treatments.

Table 1. Single contact impedances Z_{EGS} and ΔZ magnitudes at 50 Hz (merging values at $t_0 = 0$ min and $t_{30} = 30$ min) estimated on the relaxed biceps brachii muscle (no hair) of ten subjects. The model of figure 1(d), equations (2)–(4), with $I_I = 200$ nA$_{PP}$ (sine wave), 4 × 1 array with electrode surface of 5 mm^2 and inter-electrode distance IED = 10 mm as in figure 1(b) were used to estimate four values per subject (N = 40). See also figure 4.

Single contact impedance $\lvert Z_{EGS}\rvert$ at 50 Hz (kΩ)			
Skin treatment	Mean ± std. dev.	Median (2nd and 3rd interq.)	Range
No treatment (NT)	845 ± 327	861 (648–1062)	287–1465
Rubbing with ethyl alcohol (RA)	668 ± 468	655.90 (266–1054)	24–1776
Rubbing with abrasive paste (RP)	21.8 ± 17.0	15.80 (10.05–29.45)	4.90–78.20
Stripping with adhesive tape (ST)	725 ± 400	677 (426–1040)	128–1739
Impedance unbalance $\lvert\Delta Z\rvert = \lvert Z_{EGS1} - Z_{EGS2}\rvert$ at 50 Hz (kΩ)			
No treatment (NT)	150.7 ± 85.9	163.6 (70.7–208.7)	30.2–343.8
Rubbing with ethyl alcohol (RA)	276.3 ± 340	161.0 (55.1–308.1)	34.3–1318
Rubbing with abrasive paste (RP)	12.21 ± 12.80	9.35 (2.35–17.05)	0.30–51.50
Stripping with adhesive tape (ST)	278.6 ± 226	206 (126–335)	17.9–981.3

Figure 3 shows that the three treatments (RA: alcohol, RP: abrasive conductive paste and ST: stripping) reduce the impedance magnitude and (absolute value of) phase at both t_0 and t_{30}, with respect to no treatment (NT).

The mean percentage variations of impedance (at t_0 and at 50 Hz) with respect to NT are: −22.4% for RA, −97.1% for RP, −19.4% for ST. Figure 3 also shows that the mean single electrode impedance decreases with application time from t_0 to t_{30}. The impedance unbalance $\lvert\Delta Z\rvert$ can either increase or decrease depending on the subject and can vary from a few kΩ up to hundreds of kΩ (depending on the applied skin treatment; see table 1).

The tissue impedance Z_T is < 10% of the Z_{EGS} for NT, RA and ST. However, since rubbing with abrasive conductive paste significantly reduces Z_{EGS}, Z_T is no longer negligible (<30%) with respect to Z_{EGS}. For RP treatment, the percentage ratio Z_T/Z_{EGS} (%) reaches the maximum value of 33.1% (excluding the outliers). The maximum Z_T/Z_{EGS} (%) for the different skin treatment, merging t_0 and t_{30}, are respectively: 9.71% for NT, 9.42% for RA, 33.10% for RP and 7.35% for ST.

The ANOVA applied to $\lvert Z_T\rvert$, at 50 Hz, indicated that there is a decrease of Z_T from t_0 to t_{30} ($p < 0.01$) for all the treatments (see figure 4(*a*)). These changes with time should be further investigated: they may be associated to both the absorption of the conductive gel by the stratum corneum of the skin (see Discussion, in section 4) and to the approximation of the discrete model.

The ANOVA test indicated that there is an effect of time on the Z_{EGS}, for both magnitude and phase (see Discussions in section 4). A comparison between the magnitude and phase values of all the measured Z_{EGS} after each treatment, at 50 Hz, is offered by the boxplots in figures 4(*b*) and (*c*).

The ANOVA applied to the impedance unbalance $\lvert\Delta Z\rvert$, measured at 50 Hz, gave $p = 0.94$ ($F = 0.13$) for the interaction 'treatment*application time', indicating that the factors of application time and treatment do not significantly influence each other.

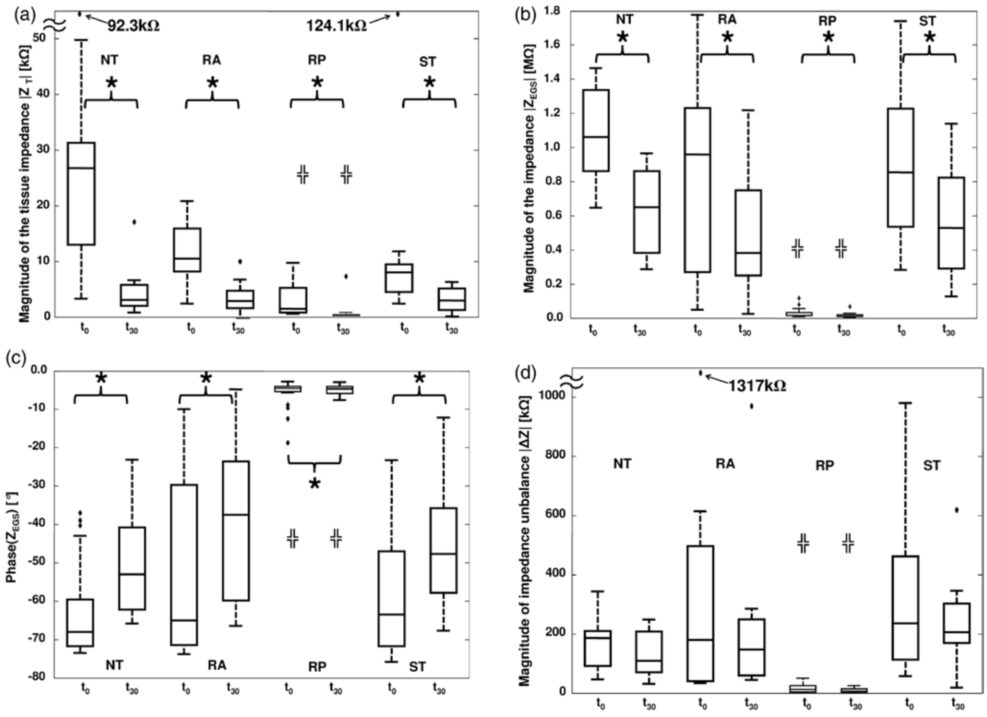

Figure 4. (*a*) Values of $|Z_T|$ at 50 Hz for all the treatments, at times t_0 and t_{30}. For each box the number of the samples is $N = 10$ (10 subjects × 1 tissue impedance). The boxes represent the 1st (q1) and the 3rd (q3) interquartiles. The dashed bars represent the minimum and the maximum values excluding the outliers. Values are drawn as outliers if they are larger than q3 + 1.5(q3 − q1) or smaller than q1 − 1.5(q3 − q1). There is a statistically significant difference ($p < 0.01$) between the values of each treatment at t_0 and t_{30} (*), as well as between abrasive paste and all the other treatments (at both t_0 and t_{30}, ‡, $p < 0.01$). See text for details. (*b*) $|Z_{EGS}|$ values for all the treatments, at times t_0 and t_{30}. For each box $N = 20$ (10 subjects × 2 impedances). There is a statistically significant difference ($p < 0.01$) between the $|Z_{EGS}|$ values (mean of $|Z_B|$ and $|Z_C|$), for each treatment, at t_0 and t_{30} (*, $p < 0.01$) and between RP and all the other treatments (at both t_0 and t_{30}, ‡, $p < 0.01$). (*c*) Phase(Z_{EGS}) values for all the treatments, at times t_0 and t_{30}. For each box $N = 20$ (10 subjects × 2 impedances). There is a statistically significant difference ($p < 0.01$) between the values (mean of phase Z_B and Z_C), for each treatment, at t_0 and t_{30} (*) and between RP and all the other treatments (at both t_0 and t_{30}, ‡, $p < 0.01$). (*d*) Values of $\Delta Z = |Z_B - Z_C|$ at 50 Hz for all treatments, at times t_0 and t_{30}. For each box $N = 10$ (10 subjects × 1 impedance unbalance). There is no statistically significant difference between the values of each treatment at t_0 and t_{30}, but the RP values are significantly different from those of the other treatments (at both t_0 and t_{30}, ‡, $p < 0.01$).

The statistical analysis also showed that:

(a) only the treatment RP produces a statistically significant reduction of $|Z_{EGS}|$, with respect to no treatment, at both t_0 and t_{30}. The other two treatments, RA and ST, produce small and not significant changes of $|Z_{EGS}|$ with respect to NT (see figure 3(*a*)).
(b) the $|Z_{EGS}|$ value (at 50 Hz) at t_{30} is significantly lower ($p < 0.01$) than the value at t_0 for all treatments, suggesting an effect of time on $|Z_{EGS}|$.
(c) the absolute phase value of Z_{EGS} at t_{30} is significantly lower than the value at t_0 for all treatments, indicating a more resistive Z_{EGS} at t_{30}. Only the treatment RP produces a statistically significant reduction of absolute phase with respect to NT ($p < 0.01$) and to the other two treatments ($p < 0.01$). RA and ST do not produce significant changes of phase of Z_{EGS} with respect to NT (see figure 3(*b*)).

(d) ΔZ at 50 Hz shows no significant difference between t_0 and t_{30}. RP results in values of ΔZ significantly lower ($p < 0.01$) than those produced by NT or other treatments (figure 4(*d*)) and is therefore the treatment leading to minimal power line interference (figure 1(*a*)).

These results are confirmed by the mixed model analysis made using the 'R' package 'lme4' (Bates *et al* 2014) with 'treatment' and 'time' as factors and 'subject' as additive scalar random effect. With this analysis, precise distributional results and *p*-values are not available, as discussed by D Bates in https://stat.ethz.ch/pipermail/r-help/2006-May/094765.html, and therefore the linear regression analysis was considered as the main analysis.

3.2. Noise measurements

The same analysis that was carried out on Z_{EGS} was applied to RMS noise values V_{RMS}, measured between electrode pairs in the band 10–1000 Hz, as shown in figure 2(*b*) for the different treatments at the two application times. For each treatment, each time, and each subject, the noise measurement was repeated six times (six electrode pairs, considered as replicate measurements). The results are shown in figure 5.

The linear regression analysis on the mean of the six RMS noise values, followed by Newman–Keuls *post hoc* tests showed a significant increase between t_0 and t_{30}, for the noise values after NT ($p < 0.01$), RA ($p < 0.01$) and RP ($p < 0.05$), but not for ST. Newman–Keuls *post hoc* showed that noise values after RP treatment are significantly lower than after the other treatments (maximum RMS noise value for RP: 3.24 μV_{RMS}, excluding the outliers).

The same analysis on the RMS noise values returned $p < 0.01$ for the interaction 'Treatment*Application time' ($N = 54$ for each treatment at each application time; 9 subjects × 6 noise RMS values between the electrode pairs; subject #1 has been excluded from the analysis because of the presence of EMG activity in the noise measurements). This indicates that the factors 'application time' and 'treatment' affect each other in determining noise level.

Attempts to estimate the variance of the noise generated by single gelled electrodes using the same approach adopted for the individual impedances (figure 1(*c*)), by replacing the impedances with the variances of the noise voltages, did not succeed. Assuming three uncorrelated noise generators, measurements of noise values between three electrode pairs led to negative estimates of the variances of the individual sources, suggesting that this simple model is inadequate for estimating the variances of the single noise generators. This issue deserves further investigation.

No consistent behaviors with time were observed between the values of Re{Z_{pair}} and the V_{RMS} values of the same pair. The Wilcoxon signed rank test was performed on the slope $\Delta V_{RMS} / \Delta Re(Z_{pair})$ between t_0 and t_{30} for each treatment. The test did not indicate significance for any of the skin treatments. There is not a statistically significant common behavior with time for the two quantities ΔV_{RMS} and ΔRe(Z_{pair}). An example of V_{RMS} versus Real(Z_{pair}) for stripping is reported in figure 6 (the other treatments showed the same behavior).

The range of values of $|Z_{EGS}|$, $|\Delta Z|$ at 50 Hz and noise V_{RMS} are summarized in tables 1 and 2. In both tables, means ±std. dev., medians, interquartiles and ranges are reported to provide information about the asymmetry of the distributions.

4. Discussion

The model described in figure 1(*d*) is particularly suitable when a 1D or 2D electrode array is used and *N*-uplets of electrodes are available. Automatic instruments based on the concepts depicted in figure 2 and equations (2)–(4) could be designed to detect Z_{EGS} (and therefore ΔZ)

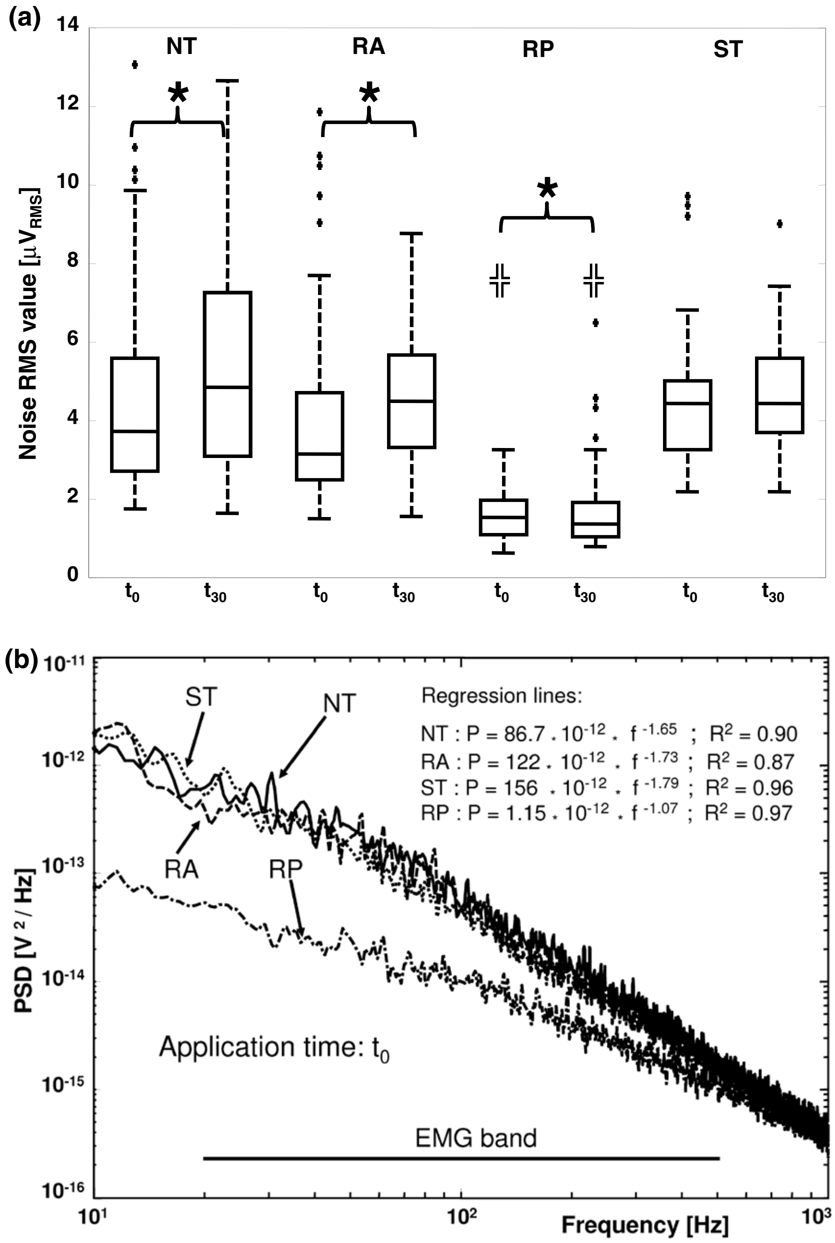

Figure 5. (*a*) Values of RMS noise (see section 2.4 and figure 2(b)) for all the treatments, at time t_0 and t_{30}. For each box $N = 54$ observations (9 subjects × 6 noise RMS values between the six electrode pairs of the electrode array; subject #1 has been excluded from the analysis because of presence of EMG activity in the noise measurements). Asterisks indicate statistically significant differences between noise amplitude values at t_0 and t_{30} ($p < 0.01$ for NT and RA, $p < 0.05$ for RP, Newman–Keuls *post hoc* tests on the linear regression analysis) and crosses indicate statistically significant differences between values after RP, compared to other treatments ($p < 0.01$). (*b*) Average power spectral densities for the four skin treatments, at t_0. Noise PSD in the band 10–1000 Hz is significantly lower for RP with respect to NT, RA and ST.

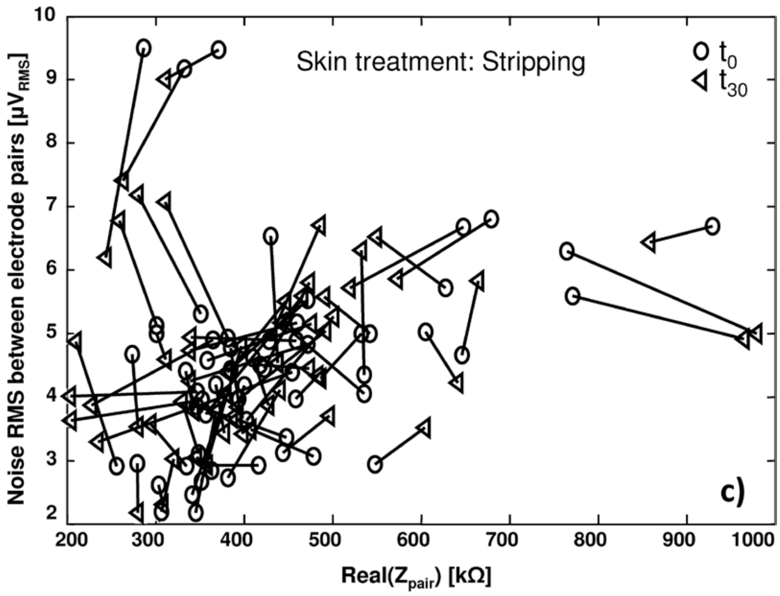

Figure 6. Example of V_{RMS} versus Re(Z_{pair}) at t_0 and t_{30} for the stripping treatment. The Wilcoxon signed rank test performed on the ratio ΔV_{RMS} / ΔRe(Z_{pair}) between t_0 and t_{30} did not show a statistically significant common behavior with time for the two quantities.

at 50 Hz, as well as noise, after the application of the array, and monitor them at specific times after application, in the case of tests of long duration. For few electrodes (mostly in ECG applications) the traditional methods, such as that of Grimnes (Grimnes 1983) would be suitable as well, but would require separate electronic circuitry.

Contrary to expectation and common practices, rubbing the skin with alcohol or stripping it with adhesive tape (five times), do not significantly lower the single impedance Z_{EGS}, nor the impedance unbalance ΔZ, nor the RMS noise V_{RMS}. In some cases, rubbing the skin with alcohol or stripping increased Z_{EGS} and ΔZ, which is opposite to the desired results. Only rubbing with abrasive paste (RP) markedly reduced Z_{EGS}, ΔZ, and noise V_{RMS}. Stripping with tape up to 12 times would likely be more effective (de Talhouet and Webster 1996) but clinically unacceptable.

Figures 3(*a*) and 4(*b*) show that the $|Z_{EGS}|$ decreases with application time from t_0 to t_{30}. This behavior of Z_{EGS} is most likely due to the absorption of the gel by the skin, possibly through the pores, which causes an increase in the skin conductivity. On the contrary, $|\Delta Z|$ can either increase or decrease in time depending on the subject.

Contrary to expectation, in many cases Z_T decreases, within 30 min, occasionally by almost an order of magnitude (figure 4(*a*)). Even assuming that the electrode conductive gel penetrates the skin and lowers the impedance of its most superficial layers (that is why Z_{EGS} decreases with time), it is not likely that the gel reaches the deeper layers and lowers the tissue impedance Z_T as much as observed. Therefore, the model depicted in figure 1(*d*) is likely too simple for the purpose of studying the tissue impedance Z_T, due to the lack of a clear boundary between Z_{EGS} and Z_T in a system that is continuous and not discrete as assumed in the proposed model. Very likely, a superficial impedance (Z_S) located in the superficial layers of the skin between two electrodes, which are close to each other, should be accounted for and investigated in future works. Z_S splits the EGS impedances in two (Z_{EGS}' and Z_{EGS}'') or

Table 2. Noise RMS values measured from electrode pairs for different skin treatments merging t_0 and t_{30}, frequency band = [10–1000] Hz. Nine subjects (subject #1 excluded due to EMG activity in noise recordings), six electrode pairs measured for each subject, at t_0 and t_{30}. ($N = 9$ subjects × 6 electrode pairs × 2 measurement times = 108). See also figure 5.

Skin treatment	Mean ± std. dev. (μV_{RMS})	Median (2nd and 3rd Interq.) (μV_{RMS})	Range (μV_{RMS})
No Treatment (NT)	4.96 ± 2.75	4.14 (2.74–6.49)	1.62–13.00
Rubbing with ethyl alcohol (RA)	4.32 ± 2.08	3.86 (2.35–5.14)	1.51–11.80
Rubbing with abrasive paste (RP)	1.67 ± 0.85	1.48 (1.07–1.96)	0.63–6.48
Stripping with adhesive tape (ST)	4.62 ± 1.48	4.44 (3.64–5.22)	2.17–9.50

more layers and forms a parallel circuit including the impedance $Z_A'' + Z_T + Z_B''$ as shown in figure 7. Progressive penetration of the conductive gel in the pores and between pores might progressively reduce Z_S as well. While this is an important issue for further investigation of the skin electrical behavior, it is not so relevant for the practical purposes addressed in this work, whose objectives focus on a new technique and on its application to assess the effect of commonly used skin treatments on Z_{EGS}, ΔZ, and noise V_{RMS}. Future work on dry electrodes, or more dense gels, will likely clarify this issue, which deserves attention as smaller and smaller inter-electrode distances will be used in EMG electrode grids.

Since the edge between Z_{EGS} and Z_T is blurred, a reduction of the estimated Z_T could be due to a reduction of Z_{EGS}'' with time, as evident from figure 7. Future improvement of the proposed procedure may be focused on a better technique for the estimation of the tissue impedance by means of a distributed impedance model. At the moment, the results concerning Z_T should be considered with caution, as a limitation of this study.

The tissue impedance is negligible with respect to the EGS impedances for all the treatments except for RP. This is due to the effectiveness of RP in lowering Z_{EGS}, thus making Z_T/Z_{EGS} higher than the other treatments.

When NT, RA or ST are applied, and the Z_{EGS} values are high, Z_T can be neglected with acceptable errors. The maximum relative errors in the estimate of Z_{EGS}, due to neglecting the relative contribution of the tissue impedance Z_T with respect to the impedance between two electrodes ($Z_{pair} = Z_{EGS1} + Z_T + Z_{EGS2}$), when the two Z_{EGS} have equal values, are: (i) 4.62% for NT, RA and ST when Z_T is 9.7% of Z_{EGS} (max measured value of Z_T/Z_{EGS} (%) in these three treatments) and (ii) 14.20% for RP when Z_T is 33.1% of Z_{EGS} (max measured value for this treatment).

Figures 4(*a*) and (*b*) show that RA, RP and ST lower the impedance magnitude and raise the (negative) impedance phase, in the considered bandwidth, at t_0 and t_{30}, with respect to NT. It seems, from figures 4(*a*) and (*b*), that a skin treatment is always preferable, with respect to the NT condition. In particular, Z_{EGS} shows a decrease of magnitude and an increase in phase (less negative) from t_0 to t_{30}, as shown in figures 4(*b*) and (*c*).

Knowing that $|Z| = \sqrt{\mathrm{real}(Z)^2 + \mathrm{imag}(Z)^2}$ and $\angle Z = \mathrm{arctg}\,[\mathrm{imag}\,(Z)\,/\,\mathrm{real}\,(Z)\,]$, we can deduce that the quantity $\mathrm{real}(Z)^2 + \mathrm{imag}(Z)^2$ decreases with time, while the imaginary part of the EGS impedance decreases (it becomes less negative) faster than the real part. This implies that the EGS impedance becomes progressively more resistive in all the treatments. Resistive behavior of an electrode impedance is generally desirable because of reduced filtering effects in the circuit depicted in figure 1(*a*) and therefore reduced waveform change, which is desirable for surface EMG decomposition (Holobar *et al* 2009).

The evolution with time of Z_{EGS} is similar to that reported in (Searle *et al* 2000, Hewson *et al* 2003), where a decrease of the impedance magnitude between large surface electrode pairs (>20 mm^2) was observed.

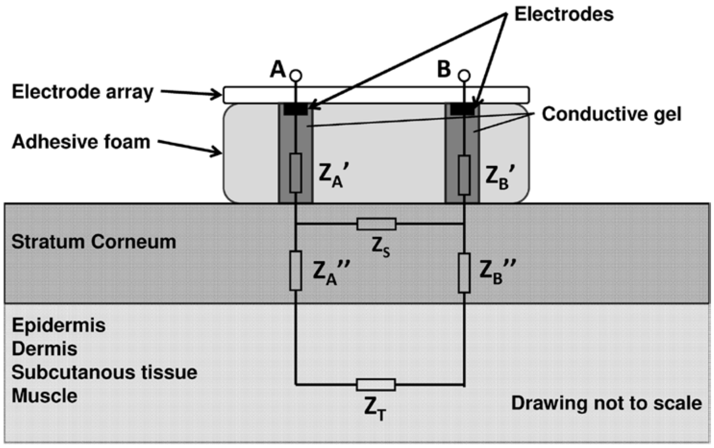

Figure 7. Impedance model between two electrodes, with small inter-electrode distance, that would explain a reduction with time of the estimated tissue impedance Z_T. Z_S is the superficial impedance between two electrodes. It is high compared to the Z_T and to the Z_{EGS} and is located on the most superficial layers of the skin. Z_S splits the EGS impedances into two parts, Z_{EGS}' and Z_{EGS}'', with $Z_{EGS}'' << Z_{EGS}'$ (Z_{EGS} represents either $Z_A = Z_A' + Z_A''$ or $Z_B = Z_B' + Z_B''$) and is in parallel with the sum of impedances $Z_A'' + Z_T + Z_B''$. A change of Z_S, Z_A'', Z_B'', due to gel penetration is interpreted by the measurement system as a change of Z_T.

The only treatment that significantly reduced $|Z_{EGS}|$ and absolute phase (Z_{EGS}), thus making the Z_{EGS} smaller and more resistive, is RP ($p < 0.01$ with respect to any of the other treatments). The statistical analysis showed that the impedance unbalance $|\Delta Z|$ after RP is significantly smaller than after any of the other treatments, at either t_0 or t_{30}.

There are no significant differences between the $|\Delta Z|$ values after NT, RA and ST at either t_0 or t_{30}. Alcohol and stripping procedures are not effective in reducing the impedance unbalance and in some cases they increase the single EGS impedance. Despite the decrease in individual $|Z_{EGS}|$ from t_0 to t_{30}, the impedance unbalance $|\Delta Z|$ does not significantly change during the same time interval so that the common mode interference that is converted into a differential voltage because of $|\Delta Z|$ (figure 1(*a*)) does not significantly change amplitude during the first 30 min of electrode application.

Linear regression analysis has been applied to the noise RMS values, measured between electrode pairs in the band 10–1000 Hz, after the different treatments at t_0 and t_{30}. A significant difference was observed, for each treatment between t_0 and t_{30} (except for ST) with higher noise values at t_{30} ($p < 0.01$ for NT, $p < 0.01$ for RA and $p < 0.05$ for RP). This is an unexpected outcome, because the noise is expected to be (at least partially) related to the real part of the electrode impedances which decreases with time. The Johnson noise attributable to the real part of the Z_{EGS} of electrode pairs, in the 10–1000 Hz bandwidth, is in the range of [0.58–4.62] μV_{RMS} for RP and [1.32–9.05] μV_{RMS} for the other treatments, while the measured noise is higher and respectively [0.63–6.48] μV_{RMS} and [1.50–13.06] μV_{RMS}. In addition, the changes with time (increase or decrease) of Re(Z_{pair}) are not associated to changes of noise amplitude for the same electrode pair. This may be related to the fact that (a) Re(Z_{pair}) is not a physical resistor (it is a mathematical object which is a function of frequency), and (b) the source of electrode noise is not (only) the Johnson noise. In agreement with the findings of Huigen *et al* (Huigen *et al* 2002) the Johnson noise due to the resistive component of the EGS interface (a) is only a fraction of the electrode noise and (b) has an erratic behavior with time. Huigen *et al* found a correlation between noise RMS values and the parallel resistive component of the impedance between the electrodes of a pair. However, the results reported in figure 6 show no relation between noise RMS values and the real part of the series impedance. Considering all the six possible electrode pairs of the model in equation (2), the overall noise RMS range of values for the best skin treatment (RP) is 0.63–6.48 μV_{RMS} (from table 2: mean ± std. dev.: 1.67 μV_{RMS} ± 0.85 μV_{RMS}). The fact that this treatment provides the lowest noise level is not due to the lower resistive component of Z_{EGS} and remains to be investigated.

4.1. Limitations of the study

There are several limitations in this study. It has been reported that electrode noise depends on the electrolytic gel used, on the subject and on the electrode location on the body (Riistama and Lekkala 2006), in addition to the electrode type and surface (Huigen *et al* 2002). In our study, the electrode arrays were placed on one specific site (biceps brachii) where there is no hair which might affect the EGS impedance. Secondly, a single gel type was used to fill the cavities of the electrodes. There is a variety of conductive gels and pastes commercially available, but their composition is usually unknown and not available from the manufacturer. The effect of different gels on the interface properties remains to be investigated.

Impedance and noise of the EGS contact have been analyzed at t_0 and t_{30} but their time course in this interval has not been investigated. Finally, the model proposed in figure 1(*d*) has shown limitations when Z_T is not very small with respect to Z_{EGS}.

5. Conclusions

A new approach is proposed to measure the complex electrode–gel–skin (EGS) impedance Z_{EGS} of single small electrodes used in arrays for HDsEMG while accounting for the tissue impedance Z_T between them. The method is based on multiple measurements of electrode pair impedances (Z_{pair}). The required hardware could be implemented with the front-end electronics of a large electrode array to perform preliminary and periodic testing during long term recordings. The new approach has been validated and applied to compare three clinically common skin treatments (RA, RP, ST) with NT and verify their effectiveness in lowering the Z_{EGS} and $|\Delta Z|$ impedances and to study the association between Re(Z_{pair}) and noise.

The main conclusions are:

(a) Among the four considered skin treatments, the one producing the lowest values of $|Z_{EGS}|$, $|\Delta Z|$ and noise V_{RMS} is RP. This treatment provides $|Z_{EGS}|$, $|\Delta Z|$ and noise V_{RMS} significantly smaller than those of the other three treatments. RA and ST (five strips) do not produce results significantly different from NT. The relative contribution of the tissue impedance Z_T to the impedance between two electrodes can be neglected for NT, RA and ST but not for RP.

(b) The Z_{EGS} becomes progressively lower and more resistive with time, while the impedance unbalance $|\Delta Z|$ does not significantly change in the first half hour of electrode application, for any of the tested treatments.

(c) Noise levels increase with time for NT and two of the tested treatments (RA and RP) but not for ST (five strips). The changes of noise amplitude in time are not related to changes of the real part of the impedance between electrode pairs, Re(Z_{pair}).

(d) Other dry or wet electrodes and electrode-gel configurations should be investigated, with the proposed approach, to clarify some of the open issues.

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