Online Skin-Electrode Contact Quality Monitoring in Wearable Devices: An EEG Application

Valentina Casadei, Roberto Ferrero
Department of Electrical Engineering & Electronics
University of Liverpool, Liverpool, UK
Valentina.Casadei@liverpool.ac.uk
Roberto.Ferrero@liverpool.ac.uk

***Abstract*—Wearable devices are becoming increasingly promising for many applications, but the accuracy of their measurements relies on good skin-electrode contact quality throughout the recording. While it is possible to assess the contact quality before the recording by measuring the impedance between electrodes, an accurate and reliable tool to monitor the contact quality during the recording is still unavailable. This is especially important for EEG recordings, because of the very low signal-to-noise ratio of EEG signals. This paper proposes an innovative approach for continuous monitoring of the skin-electrode contact quality, based on the measurement of the line-frequency radiated interference, always present in indoor environments. Differently from other existing methods that measure the absolute interference power, the proposed method uses the relative 50 Hz power normalized to a reference channel, to detect contact variations in individual channels that may occur during the recording, due to possible electrode displacement, gel dehydration or variations in contact pressure. The proposed method has been tested on a commercial wearable EEG device and shows some advantages compared to the signal quality index calculated by the device, in terms of sensitivity to small impedance differences and robustness in presence of muscular artifacts and environmental changes in the source of interference.**

***Index Terms*—Wearable sensors, Biomedical measurement, Biomedical monitoring, Biomedical electrodes, Bioimpedance, Electroencephalography**

I. Introduction

Wearable devices have lately become of increasing importance for healthcare monitoring and research purposes, by wirelessly recording biomedical measurements, such as electrocardiography (ECG), skin conductance, electromyography (EMG), electroencephalography (EEG). The proliferation of the wearables is due to their simplicity in being worn and used, providing a cheaper alternative at the same time, without the assistance of an expert [1]. However, the accuracy of the measurement results remains a concern, especially in relation to the recorded signal quality, which is strongly correlated to the skin-electrode contact quality [2], [3]. The latter can be affected by several factors: 1) the type of electrode or wearable sensor used (dry, gel based, solid gel, textile, etc.) [4], [5], 2) electrode misplacement (inexperience of the user), 3) skin preparation [6] and 4) deterioration of the contact over time, especially in very long acquisitions.

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According to the current state-of-the-art [7], the most accurate assessment of the skin-electrode contact quality can be achieved by measuring the impedance between two electrodes (one of them being typically the reference electrode). However, the impedance measurement requires the injection of a small current, usually in the frequency band of the signal to be recorded, which is not compatible with the signal acquisition; for this reason, the impedance measurement is usually performed before (and/or immediately after) the recording, but not during it. While this is useful to detect the quality of the initial contact, this method is unable to detect any deterioration of the contact quality during the acquisition. Monitoring the quality of the contact during the whole signal acquisition would be of great interest, since the contact is likely to be influenced by pressure variations [8], head movement, drying gel, etc. A possible solution to measure the impedance during the acquisition was suggested in [9], but the measurement was performed outside the signal bandwidth; moreover, this solution may not be always technically feasible, as it requires appropriate hardware. Other solutions to assess the skin-electrode contact during the recording are therefore desirable.

Monitoring the signal quality is particularly important in EEG devices, specifically for engineering applications, such as Brain-Computer Interfaces (BCI), because the EEG Signal-to-Noise ratio is already very small and a very large impedance would further decrease it, compromising the processing. Some research-grade EEG devices (e.g. the Neuroelectrics Enobio [10]) provide the user with a color-based quality index (QI), which acts more as a guidance than a proper quality assessment and it is suggested to be used together with continuous visual inspection of the raw signals during the recording, as explained by the manufacturer; moreover, the quality index refers to the overall signal quality, including artifacts, not only the contact quality. Other research-grade EEG devices (e.g. the ANT Neuro Eego [11]) provide the impedance measurement before and after the acquisition, but no other quality assessment during the recording.

In this scenario, the main limitation appears to be the lack of a unique, accurate and reliable index for monitoring the biosignal quality. The most commonly used indicator of signal quality is the line-frequency noise (at 50 Hz or 60 Hz, depending on the country), which is always present in indoor acquisitions, because of radiated interference from nearby

mains circuits [12], [13]. Its amplitude is strongly affected by the skin-electrode impedance, whose higher values lead to higher noise amplitudes. However, the noise amplitude also depends on the source of interference in the acquisition environment, which is likely to change in different tests and may also vary during each recording. Therefore, the line noise cannot be accurately used as an absolute indicator of contact or signal quality. To address this limitation, this paper proposes an alternative use of the line noise, not to evaluate the absolute contact quality, but to detect relative variations between channels over time. Together with an initial impedance measurement, this method can provide an accurate and robust assessment of any skin-electrode contact degradation during the acquisition.

II. EEG SIGNAL QUALITY ASSESSMENT

The unavailability of impedance measurements, combined with the absence of a standard index to assess the quality of the raw EEG signal during the acquisition, have led different researchers and manufacturers to define different metrics to be used either alone or together. The main metrics proposed in the literature are summarized in the following [14] and are calculated on time windows of one or a few seconds:

- Line Noise: power (or RMS value) of the signal in a narrow band around the line frequency (50 Hz or 60 Hz). As explained before, this index is strongly correlated with the skin-electrode impedance, but it also depends on the source of interference, which is usually neither known nor constant.
- Offset: mean value of the waveform within the window. A large offset in the signal is often a good indicator of a skin-electrode contact with high impedance; however, the relationship between them is not deterministic and a large impedance does not necessarily lead to a large offset. Moreover, the offset may appear with a significant delay after the contact deteriorates, and it may change over time.
- Drift of the signal: it can be seen as a continuous variation of the offset, which may indicate a large contact impedance, as explained above. However, in wet electrodes, a drift may also appear during the gel stabilization transient and it does not necessarily imply a poor skin-electrode contact.
- EEG signal amplitude: power (or RMS value) of the signal in the main EEG frequency band (e.g. 1-40 Hz). This index can well detect artifacts whose amplitudes are much higher than typical EEG signals.
- Overall RMS value: amplitude of the whole signal in the window. This index has a similar meaning as the previous one, but it is more sensitive to lower-frequency components and possibly high-frequency noise, if not filtered.
- Maximum gradient: it is the largest difference between adjacent samples within the window. It is usually used to detect spike-like artifacts, e.g. setting 10 μV/ms as a threshold.
- Zero-crossing rate (ZCR): rate at which the signal changes its sign. It is an indicator of the dominant frequency of the signal and it can show high values when the signal contains high-frequency noise or artifacts. However, this index can be heavily affected by low-frequency components and offset.
- Kurtosis: it is the standard statistical measure of the heaviness of the tails of a distribution of samples. Large values of the kurtosis reveal significant outliers in the distribution, which could indicate the presence of some large artifacts with short duration [15].

Among them, the first three metrics are related to the contact quality, while the others mainly reveal artifacts, whose presence does not necessarily imply a poor skin-electrode contact (e.g. they could be muscular artifacts). Since each of the indices above has important limitations, several of them are often used together to achieve a more reliable assessment of the EEG signal quality. The following subsections report two examples, from a commercial product and a research study, respectively.

A. Neuroelectrics® Enobio

The Neuroelectrics Enobio EEG device is a medical grade wireless device for the real-time analysis of brain signals. It has been used in several research studies, including medical applications (sleep monitoring and epilepsy [16]), emotional state assessment and BCI applications [17], which are the main intended applications.

The EEG quality measurement during the acquisition is done via a quality index (QI) [18], which is computed within a time window of 2 s:

$$QI = \tanh \sqrt{\left(\frac{Off}{W_{off}}\right)^2 + \left(\frac{EEG}{W_{EEG}}\right)^2 + \left(\frac{LN}{W_{LN}}\right)^2} \quad (1)$$

Where Off is the signal offset, EEG is the rms value within the [1-40] Hz band, LN is the rms value in the [49-51] Hz range (or [59-61] Hz), and their respective weights are: $W_{off} = 280$ mV, $W_{EEG} = 250$ μV and $W_{LN} = 100$ μV. The drift is also computed, but not included in the equation, because it has a high inter-subject variability. The signal quality is then subdivided by a color-based code, defining three ranges: green (QI: $0.0 - 0.5$), amber (QI: $0.5 - 0.8$) and red (QI: $0.8 - 1$).

Even though the QI can work as a general signal quality indicator, the manufacturer describes it as a guidance, not to be considered strictly accurate. In addition, it acts more as an artifact detector rather than a signal quality index.

B. Offline signal quality assessment

Not all the commercial devices provide indices for the signal quality check. Therefore, the assessment of the recorded data can be done by implementing an algorithm for their analysis offline. In [14], an index for the automated quality assessment is developed by implementing six of the EEG metrics described above: EEG and line noise signals, RMS amplitude, maximum gradient, ZCR and the kurtosis. Each

metric is calculated for segments of the EEG signal, by using a sliding window of 1 s. The signal quality is assessed on each window through the Z-scoring, comparing the average value of all the metrics to the mean value and standard deviation of a normative database of simulated EEG cleaned from artifacts. An increase in the score corresponds to a worse signal quality. However, the metrics are not totally independent of each other (for example the signal amplitude and the overall RMS value) which may affect the final score.

III. Proposed method

A wireless device is expected to be worn for long periods of time and to be used in a home environment. Hence, extracting as much information as possible from the raw signals is essential for a correct analysis of the results [12]. An approach for the continuous assessment of the skin-electrode contact quality during EEG recording is proposed, to be used in addition to the impedance measurement before the recording. The proposed method exploits the line-frequency radiated interference arising from the parasitic capacitive coupling with surrounding power circuits, which is always present in home or other indoor environments and is a common feature in all wearable recordings [13]. This interference is usually removed through the application of a notch filter, but it can provide important information regarding the quality of the signal.

As the interference power may change because of changes in the power flow in the nearby circuits or movement of the user with respect to those circuits, the absolute value of the measured power over time may not be meaningful. On the contrary, the relative power with respect to a chosen channel (or the average of all channels) is not expected to be affected by those environmental changes, because they will have similar effects on all channels; for this reason, it has been chosen for the proposed method. The average power of 50 Hz interference is calculated for each differential raw signal over a time window of 2 s, by applying the Fourier Transform and calculating the power spectrum in the range [49.5-50.5] Hz, where the line frequency is expected to lie. The resulting power values for all channels are divided by the power of a channel chosen as reference. This should be a channel with a good quality contact, which is less likely to deteriorate over time (e.g. a mid-line central) because of good pressure and limited influence of head movement. Alternatively, the average of all channels can be used as reference, but this works better for large number of channels.

It is worth noting that the chosen time window (2 s) is long enough to prevent significant leakage errors caused by the lowest-frequency components of the raw EEG signal. A high-pass filter could be used to attenuate those low-frequency components, but it is not recommended in this application because it may introduce delays and possibly long transients, with little benefit in terms of the 50 Hz power calculation. Similarly, within such short interval, the use of a non-rectangular window to decrease the spectral leakage is not recommended either, since it would worsen the frequency resolution and may therefore worsen the accuracy of the 50 Hz power estimation instead of improving it. However, if a longer window is used (several seconds or tens of seconds), then a non-rectangular window may be convenient. In this paper, a 2 s window is used for direct comparison with the QI calculation implemented in Neuroelectrics Enobio, but a longer window may be acceptable in most applications, as the skin-electrode contact quality is not expected to change significantly on such a short timescale.

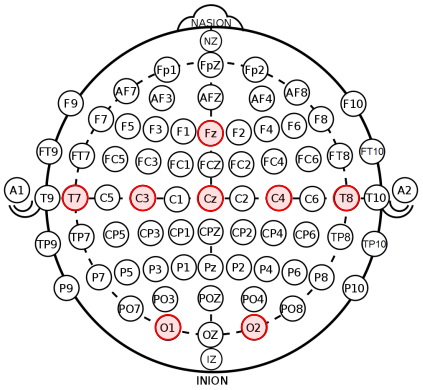

Fig. 1. Electrodes placement according to the 10-20 system. The highlighted positions are the ones considered for the tests.

IV. Experimental validation

A. Experimental setup

The proposed method has been validated using Neuroelectrics Enobio-8 on two healthy volunteers: a man with thin and short hair, and a woman with long and dense hair. On both, eight gel-based EEG electrodes were placed in Cz, Fz, T7, T8, C3, C4, O1, O2 as shown in Fig. 1, referenced to a ground electrode on the ear. All signals are sampled with 500 Hz sampling frequency and 24 bit resolution (50 nV), and the acquisition unit has a bandwidth from 0 to 125 Hz. The software allows the optional implementation of a notch filter to remove the 50 Hz noise, which has not been used in these tests. Enobio also includes a 3-axis accelerometer, which is used to record head movements. Although the signals from all channels are acquired referenced to the ground electrode, it is common practice to analyze differential signals, calculated with respect to a chosen reference electrode (here Fz). The 50 Hz noise power for each differential signal is then compared to the power of the differential signal Cz (calculated in the same way).

In all tests, the two reference electrodes (Fz and Cz) and the right-side electrodes (T8, C4 and O2) were positioned following the best practice: the skin was prepared by using abrasive paste and a large amount of gel was inserted into the electrode to ensure a good-quality contact. On the contrary, the left-side electrodes (T7, C3 and O1) were used either without preparing the skin or with very limited amount of gel, as described in more detail below, in order to purposely create a worse contact. The impedance between each electrode and Fz was measured (at 30 Hz) by the Digitimer 175 Impedance Meter, before each acquisition.

The experimental tests reported in the following subsections are aimed at investigating the effect of different contact quality, as well as the effect of head movement and of variations in the user's position with respect to the sources of interference. For all tests, the results from the proposed method are compared to Enobio's QI calculated according to (1). It should be noted that the actual QI values calculated by Enobio could not be saved together with the recorded signals, so they were re-calculated offline; therefore, there may be small differences between the QI values reported in this paper and those calculated by Enobio, but those differences are expected to be negligible. As the tests lead to the same conclusions for the two subjects, figures and detailed discussion of results are reported only for the male subject, while the numerical results for the female subject are only listed in Table I.

B. Skin preparation and gel amount

The purpose of these tests is to show how different contact impedances are reflected into different values of the relative 50 Hz power. In the first test, all electrodes had the same gel amount, but the left-side electrodes (T7, C3 and O1) were mounted without preparing the skin. This led to a very large impedance difference between the prepared and non-prepared channels, always measured with respect to Fz: for all prepared channels, the impedance was in the range from 0.5 to 2 kΩ, whereas for all non-prepared channels, the impedance was higher than 20 kΩ, so at least ten times higher than the prepared channels.

Fig. 2 reports the results of the proposed method, compared to Enobio's QI, for C3 and C4, in rest conditions (i.e., no movement and no tasks undertaken). The relative 50 Hz power of C4 is very close to 1 ($10^{0.14\pm0.28}$), indicating a good contact as expected, because C4 was prepared in the same way as the reference channel Cz. On the contrary, the relative power of C3 is much higher ($10^{3.17\pm0.27}$), showing that the proposed method is very sensitive to the contact impedance difference caused by no skin preparation. In this case, also Enobio's QI shows a significant difference between the two channels (0.03 ± 0.01 for C4 and 0.21 ± 0.01 for C3), although they both are still classified as good (green), according to Enobio's color-coded QI ranges (QI<0.5).

In the second test, reported in Fig. 3, the skin was prepared for all channels, but the left-side electrodes (T7, C3 and O1) were mounted with a much smaller amount of gel; their impedance was between 3 and 5 kΩ, slightly higher than the value of the right-side channels (between 0.5 and 2 kΩ, as before). Despite the smaller impedance difference compared to the previous test, the proposed method can still detect it between C3 and C4 ($10^{1.37\pm0.24}$ and $10^{0.15\pm0.30}$ respectively), whereas Enobio's QI shows no difference at all (0.21 ± 0.01 for both). It is worth noting that the QI values for both channels are now similar to the values of C3 in the previous test, whereas C4 showed smaller values in the previous test. Since the condition of C4 has remained the same in both tests (as confirmed also by the same impedance measurement), this result reveals how the QI value is affected by other factors

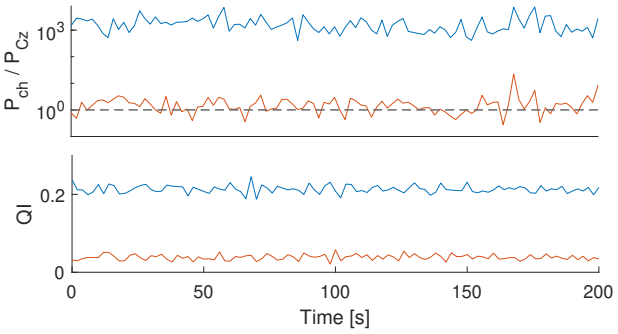

Fig. 2. Relative 50 Hz power (top) and Enobio's QI (bottom) for C3 (blue) and C4 (red), measured with skin preparation for C4 and no skin preparation for C3. The black dashed line in the top plot indicates the value 1, i.e. the expected value for a good contact.

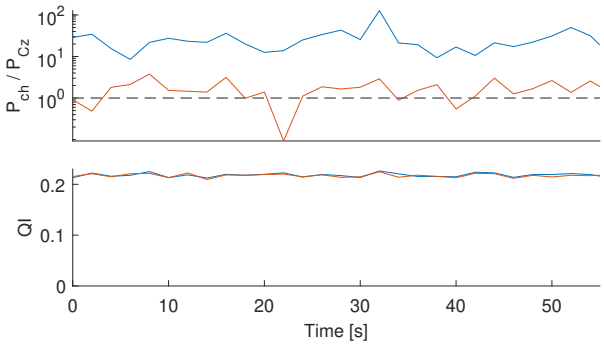

Fig. 3. Relative 50 Hz power (top) and Enobio's QI (bottom) for C3 (blue) and C4 (red), measured with large gel amount for C4 and small gel amount for C3. The black dashed line in the top plot indicates the value 1, i.e. the expected value for a good contact.

(in this case, the presence of an offset), not directly related to the skin-electrode contact quality, and therefore it is a less reliable indicator of contact quality. Additionally, the results of the proposed method are affected by significant noise because of the chosen window length of 2 s, but a longer time window would decrease it, leading to a more evident difference between the two channels.

It is worth noting that the results from the female subject (Table I) show a smaller difference in the contact quality between C3 and C4, because the long and dense hair gave rise to a larger impedance (higher than 20 kΩ) even in the good channels. Nevertheless, a difference between channels with and without skin preparation can still be revealed by the proposed method, whereas the different gel amount did not produce a significant difference in the results.

C. Head movement

Head movement is one of the most common causes of artifacts in EEG signals, especially in occipital channels, which are very sensitive to muscular activity of the neck. The presence of muscular artifacts in the EEG signal, however,

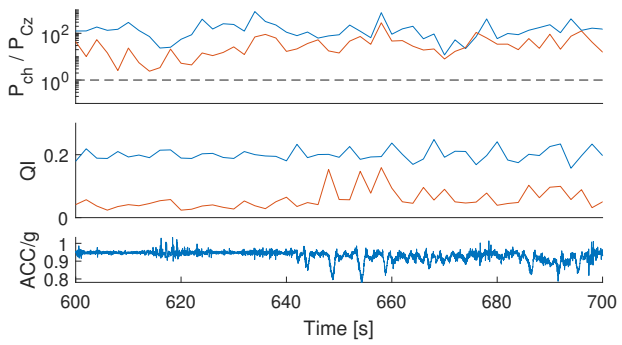

Fig. 4. Relative 50 Hz power (top) and Enobio's QI (middle) for O1 (blue) and O2 (red), measured before and during head movement. The bottom plot shows the acceleration measured by Enobio's internal accelerometer.

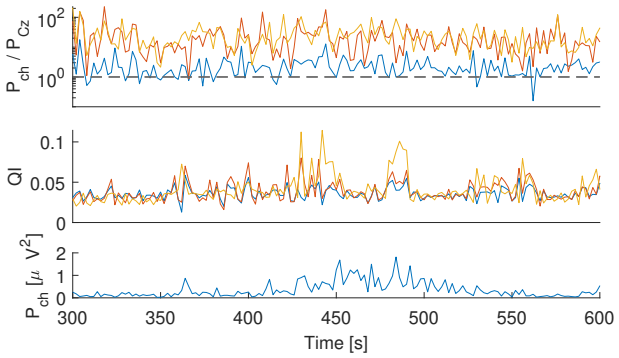

Fig. 5. Relative 50 Hz power (top) and Enobio's QI (middle) for C4 (blue), T8 (red) and O2 (yellow), measured before and during exposure to higher interference. The bottom plot shows the absolute 50 Hz power for O2, to indicate where exposure to higher interference occurred.

does not necessarily imply a worse contact quality. It is therefore important to verify that the proposed method for the assessment of the skin-electrode contact quality is not sensitive to head movement artifacts.

For this test, O1 and O2 have been selected for the results reported in Fig. 4, since they are the closest ones to the source of muscular artifacts (neck). The test conditions are the same as for the first test reported in the previous subsection, i.e. O1 (left side) is a non-prepared channel, while O2 (right side) is prepared; this condition was chosen to verify the performance of the proposed method in the two most extreme cases, with very high and very low contact impedance, respectively. In the first part of the test, the head was kept as still as possible, then it was repeatedly moved with random patterns, as can be seen from the acceleration measurement, also reported in Fig. 4.

For O2, the measured relative 50 Hz power is $10^{1.17\pm0.47}$ before the head movement (from 600 to 640 s) and $10^{1.57\pm0.32}$ during the movement (from 650 to 700 s); for O1, it is $10^{2.15\pm0.36}$ and $10^{2.04\pm0.35}$, respectively. Taking the standard deviations into account, both channels show no significant difference in the 50 Hz power before and during the movement, confirming that this indicator is not sensitive to artifacts created by the head movement.

Enobio's QI shows no significant difference before and during the movement either: for O2, its values are 0.04 ± 0.01 before the movement and 0.07 ± 0.03 during the movement; for O1, they are 0.19 ± 0.01 and 0.20 ± 0.02, respectively. However, the QI values become more noisy (larger standard deviation) during the movement and, in particular, O2 shows well visible peaks in the QI values corresponding to the largest acceleration peaks. This is caused by the fact that the QI calculation includes the signal RMS value in the [1-40] Hz band, which is affected by muscular artifacts. So, Enobio's index is potentially sensitive to muscular artifacts, although their effect may be small and only visible when the QI baseline is very low.

D. Interference source variation

The purpose of this last test is to show that the proposed use of the relative 50 Hz power is not affected by environmental changes in the source of interference, which may be caused either by changes in the power flow in the nearby mains circuits or by the user's movement with respect to those circuits. Such changes will affect the 50 Hz power measured in each channel, but the effect will be very similar on channels with the same contact impedance and therefore it will cancel out when considering the relative power normalized with respect to a reference channel, as proposed in this paper.

This is illustrated in Fig. 5, where the results from the right-side channels (C4, T8 and O2) are reported, chosen because they were prepared in the same way as the reference electrode Cz, with skin preparation and a large amount of gel. During the recording, the subject moved closer to mains-powered equipment for a short time; this caused a significant increase in the 50 Hz power in each channel, as illustrated in the bottom plot of Fig. 5 for O2 only, as an example. Despite this, the relative powers for all channels showed no significant changes: the values before (from 300 to 420 s) and during the exposure to higher interference (from 440 to 550 s) are, respectively, $10^{0.28\pm0.37}$ and $10^{0.33\pm0.27}$ for C4, $10^{1.18\pm0.48}$ and $10^{1.26\pm0.39}$ for T8, and $10^{1.28\pm0.39}$ and $10^{1.41\pm0.29}$ for O2. The slightly lower values for C4 compared to the other two channels are likely to be caused by a larger contact pressure created by the cap on that electrode, showing that the proposed method can also distinguish between small contact quality differences caused by different contact pressure.

In terms of average values, Enobio's QI does not show significant differences either, between before and during exposure to higher interference: 0.03 ± 0.01 vs. 0.03 ± 0.01 for C4, 0.03 ± 0.01 vs. 0.04 ± 0.01 for T8, and 0.02 ± 0.01 vs. 0.04 ± 0.02 for O2. However, during the exposure to higher interference, there are well visible peaks in the QI values, especially for O2 and, to a smaller extent, T8, showing that Enobio's QI may be more sensitive to environmental variations in the 50 Hz power, in agreement with the fact that

its calculation is based on the absolute 50 Hz power of the channel, not normalized to another one.

TABLE I
RESULTS OF THE FEMALE SUBJECT, OBTAINED WITH THE SAME PROCEDURE AS DESCRIBED IN SEC. IV

Channel	P_{ch}/P_{Cz}	QI_{Enobio}
Skin preparation		
C_3	$10^{1.18\pm0.07}$	0.16 ± 0.01
C_4	$10^{-0.18\pm0.14}$	0.03 ± 0.01
Gel amount		
C_3	$10^{-0.40\pm0.21}$	0.12 ± 0.02
C_4	$10^{-0.18\pm0.20}$	0.12 ± 0.02
Head movement: pre-movement (left), movement (right)		
O_1	$10^{2.98\pm0.11}$ $10^{2.80\pm0.23}$	0.55 ± 0.04 0.53 ± 0.18
O_2	$10^{0.00\pm0.15}$ $10^{0.90\pm0.36}$	0.26 ± 0.03 0.32 ± 0.13
Interference variation: exposure (left), post-exposure (right)		
C_4	$10^{-0.24\pm0.50}$ $10^{0.31\pm0.38}$	0.06 ± 0.08 0.05 ± 0.03
T_8	$10^{0.38\pm0.78}$ $10^{0.63\pm0.53}$	0.38 ± 0.08 0.38 ± 0.03
O_2	$10^{0.87\pm0.56}$ $10^{1.22\pm0.38}$	0.27 ± 0.07 0.27 ± 0.03

V. CONCLUSIONS

This paper proposed a method for continuous monitoring of skin-electrode contact quality during EEG acquisitions from wearable devices, to be used preferably in combination with an initial impedance measurement before the recording. Differently from existing signal quality indices implemented in some commercial devices, which detect also the presence of artifacts in the signal, the proposed method is intended to detect only degradation of the contact quality over time, which is likely to occur with wearable devices because of electrode (cap) displacement, gel dehydration or changes in contact pressure.

Similarly to other existing methods, the proposed approach measures the line frequency noise caused by radiated electromagnetic interference with nearby mains-powered circuits, but instead of using the absolute power measurement, it monitors the relative 50 Hz power of each channel, normalized to a reference channel (or the average of all channels). In this way, the proposed contact quality indicator is not sensitive to environmental variations in the source of interference, which would affect all channels in similar ways, but it is very sensitive to small contact impedance variations in a single channel.

The proposed method has been successfully tested using the Neuroelectrics Enobio-8 wearable EEG and it has been been compared to the signal quality index (QI) provided by Enobio. Experimental results on two healthy volunteers confirmed that the relative 50 Hz power measurement can detect contact impedance differences caused by no skin preparation, different gel amount and different contact pressure, while it remains largely unaffected by muscular artifacts caused by head movement and environmental variations in the source of 50 Hz interference. Compared to Enobio's QI, the proposed method is more sensitive to small contact impedance variations and may be less sensitive to artifacts and to the absolute value of the 50 Hz power, at least in some conditions. Therefore, it is a promising tool for continuous monitoring of the skin-electrode contact impedance, to detect any contact degradation that may occur during the acquisition, after the initial recommended check by impedance measurement.

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