35th Annual International Conference of the IEEE EMBS
Osaka, Japan, 3 - 7 July, 2013

Investigating the Impact of Force and Movements on Impedance Magnitude and EEG

Vojkan Mihajlović[1] Haoxuan Li[2] Bernard Grundlehner[1] Julien Penders[1] and Alfred C. Schouten[2,3]

Abstract— **The success of applying dry sensor technology in measuring electroencephalogram (EEG) signals will have a significant impact on a wider adoption of brain activity monitoring in ambulatory as well as real life solutions. The presence of motion artifacts is the major obstacle in applying dry sensors for long-term EEG monitoring. In this paper we assess the impact of external forces applied on a dry EEG electrode as well as the impact of head and body movements on the electrode-tissue contact impedance and the EEG signal. The data collection method and the preliminary correlation analysis are presented. The analysis demonstrates that the impedance magnitude and EEG changes are highly correlated when artifacts are induced by the application of force or head and body movements, only in case these artifacts are short (less than 3s) and exhibit regular pattern. The correlation between the EEG and impedance magnitude is lower for longer artifact segments, especially the ones containing artifacts with irregular movements or large variations in the applied force. This indicates a time-dependent, non-linear relation between the artifact-related phenomena, impedance magnitude, and EEG.**

I. INTRODUCTION

Recent technological developments in the area of noninvasive monitoring of electrical activity of the brain (electroencephalography, EEG) focus on sensors that do not use conductive gel and skin preparation but instead a so called dry-contact electrodes [1]. While dry electrode systems demonstrate short setup time and increased user comfort they are faced with decreased signal quality compared to ambulatory EEG systems [2], [3]. To go beyond gaming as an application domain and to be considered as an alternative to ambulatory EEG systems or gel-based lifestyle solutions, such dry electrode systems need to incorporate sophisticated signal processing algorithms. Among the most devastating disturbances present in the EEG signal recorded with dry electrodes is the impact of motion artifacts [4]. Therefore, handling motion artifacts is the most important challenge that dry electrode systems should address.

Given that most studies with dry electrode systems are done in controlled environments where users are instructed to avoid movements during the recording procedure, only a handful of authors discus motion artifacts. While the option of avoiding these might be acceptable in the laboratory setting, in real life situations and in ambulatory recordings this is an unrealistic scenario. In such situations, extracting movement patterns and using them for the artifact removal, as in e.g., [5], might not be feasible due to irregularity of the motion patterns. This is why we focus on characterizing the effects of human motion on EEG. We believe that relying solely on human EEG signal will not suffice in estimating the impact of artifacts, hence monitoring phenomena that are related to movements is required.

Since motion artifacts are mainly caused by changes on the electrode-skin contact interface, continuous monitoring of electrode-tissue contact impedance is of great importance in estimating the EEG signal quality and removing motion components from the EEG signal [6]. The objective of this paper is to investigate the impact of force applied on the electrode and natural movements on electrode-tissue contact impedance and EEG, measured with the wireless EEG system developed within imec [7]. The evaluation builds on the experience from our preliminary study on the impact of force on electrode-tissue contact impedance [8]. A comprehensive measurement setup and evaluation protocol is presented, aimed at simultaneous extraction of EEG and impedance. The elementary analysis of obtained signals in terms of absolute values, variations, and correlation between signals is presented, and the outcome of the analysis is discussed.

II. METHODS

This section discusses the evaluation setup, protocol, and data analysis. The evaluation setup presented in this paper is a subset of a larger setup which details are beyond the scope of this paper.

A. Evaluation Setup

The measurement setup consists of the following (as illustrated in Figure 1):

- A BioPac EL120 dry Ag/AgCl measurement electrode containing rigid pins that can penetrate the hair, positioned at Cz location (according to the International 10-20 System for EEG measurements)
- Two Ag/AgCl cup electrodes filled with conductive gel, positioned behind left (reference electrode) and right (patient bias electrode) ear
- A rigid headset for holding the dry electrode in place
- An EEG signal acquisition system for measuring impedance and EEG

One of the important features of the EEG measurement system is that it can continuously monitor electrode-tissue

[1]V. Mihajlović, B. Grundlehner, and J. Penders are with Holst Centre / imec-nl, 5656 AE Eindhoven, The Netherlands {vojkan.mihajlovic, bernard.grundlehner, julien.penders} at imec-nl.nl

[2]H. Li and A. C. Schouten are with the Faculty of Mechanical Engineering, Delft University of Technology, 2628 CD Delft, The Netherlands {a.c.schouten, h.li} at tudelft.nl

[3]A. C. Schouten is also with the Faculty of Engineering Technology, University of Twente, Enschede, The Netherlands

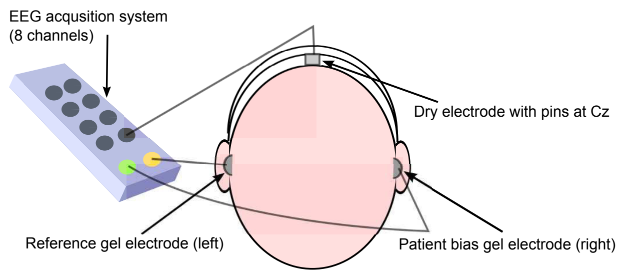

Fig. 1. Illustration of the evaluation setup and its components.

impedance along with the EEG data [7]. To perform the impedance measurement, the acquisition system includes electric current generation modules for each electrode, within the active electrode front end. This active electrode front end can induce a square wave current at the frequency of 1024Hz. The amplitude of the square wave (I_d) can vary between 10nA and 2mA and for the evaluation it was set to a value of 50nA. The measured voltage on the scalp due to impedance is buffered at the active electrode front end and amplified 100 times in the back-end stage. The induced 1024Hz signal is than demodulated and the first harmonic of the square wave is amplified. The impedance magnitude is measured by demodulating in-phase (Z^i) and quadratic component (Z^q) of the impedance signal. The EEG signal shares most of the same path in the amplification (without the demodulation step) as the impedance signal. Both, impedance and EEG signal are digitized using a 12bit ADC. The sampling rate at the ADC was set to 1024Hz. Based on the output of the EEG system impedance magnitude and EEG are calculated. The acquisition and the visualization of the EEG and impedance magnitude values in real-time is performed using internally developed Matlab functions, which also store the recorded data for offline analysis.

B. Evaluation protocol

We included 11 participants (8 male and 3 female, between 22 and 36 years of age) in the evaluation. They all signed informed consent before participating in the experiment. During the experiment they were seated in a comfortable chair and the setup was mounted on their head. The experimenter ensured that the quality of the signals was at the desired level by visual inspection and by readjusting the headset if required. During the evaluation the experimenter rendered the force by pressing the headset with a firm object while the subject was instructed beforehand on how to mimic natural movements, consisting of head movement in the sagittal and coronal plane, standing up from a chair, and walking and jumping on the spot. The evaluation consisted of the following eight sessions, such that each session started with 90s where no force was applied at the Cz electrode and no movements were performed:

- *Continuous force:* Periods of 60s of different constant force applied on Cz electrode were interspersed with 60s of no force application. The forces applied were maintained at around 1.5, 2.5, and 4N in each of the 60s segments.
- *Repetitive force:* Periods of 60s of different repetitive force applied on Cz electrode were interspersed with 60s of no force application. The periods of force application were 3, 5, and 10s in each of the segments. The force was maintained at around 2N.
- *Impact force:* Period of 180s was recorded where different punch forces were applied on Cz electrode. The force were rendered by punching the headset every 3 to 5s, such that the peak forces applied were in the range of 1 to 5N.
- *Head movement in sagittal plane:* The subject was asked to move his head in a sagittal plane (a movement that resembles head nodding) for 2s followed by 2s of no head movement. In total, about 10 of such head movements were recorded.
- *Head movement in coronal plane:* The subject was asked to move his head in a coronal plane (a movement that resembles left-right head tilting) for 5s followed by 2s of no head movement. In total about 10 of such head movements were recorded.
- *Stand up and sit down:* The subject was asked to stand up from a chair without any head movement followed by 5s of standing on the spot. Then the subject was asked to sit down without any head movement followed by 5s of sitting on the chair. In total about 10 of such head movements were recorded.
- *Walking on the spot:* The subject was asked to walk slowly (1 step per second), at a normal pace (2 steps per second), and fast (3 steps per second) on the spot, continuously for 60s for each of the walking paces. Between each walking segment 60s break was applied.
- *Jumping on the spot:* The subject was asked to jump from the standing position. Every jump was followed by the period of 5s without any head movement. In total about 10 of such jumps were recorded.

After these eight sessions the setup was removed from the participants's head and he/she was provided with an opportunity to give feedback about the experiment. The total duration of the experiment was between 60 and 90 minutes per participant.

C. Data Analysis

The analysis of recorded data was done using internally developed Matlab functions. To estimate the impact of applied force and motion artifacts on impedance magnitude and EEG, essential properties of the signals are first extracted. The 20Hz low pass filter was applied to both signals and they are resampled at 128Hz. We report mean value and variations in impedance magnitude and EEG in terms of the difference between the maximum and minimum values and the standard deviation across all eight different sessions and across participants. To investigate to what extent changes induced by motion artifacts, reflected in the skin-electrode contact surface changes, can impact changes in the EEG, a correlation analysis using Pearson product moment correlation coefficient between EEG and impedance magnitude is performed. The correlation analysis is performed on segments that encompass the large changes in the signal induced due to either force application and release or due to movement activity. We selected shorter segments to avoid artifacts other than the ones induced. Such artifacts are often present at the beginning or towards the end of the recording. The duration of the selected segments was 3s for continuous and impact force, nodding, standing-up/sitting-down and jumping sessions, 6s for repetitive force with 3s period and head tilting sessions, 10s for repetitive force with 5s period, 15s for repetitive force with 10s period, and 80s for walking sessions. The correlation values per participant and per subject are reported.

III. RESULTS

An example of the obtained EEG and impedance magnitude values is given in Figure 2. The figure illustrates how head movements in the coronal plain (tilting the head) results in clear changes in the impedance magnitude as well as EEG. Clearly, 13 segments of head movement can be identified. The figure shows that the changes in the EEG signal due to motion have similar shape as the changes in the impedance magnitude signal. Similar observations can be made for other types of motion or force applied, and across many subjects. This indicates that the two signals might be highly correlated.

To compare the impact of different movements and forces applied have on two signals we report properties of the impedance magnitude and EEG signals in Table I. The table shows that the mean impedance magnitude stays in the range of $20k\Omega$ to $80k\Omega$. There is a significant difference among subjects and across different conditions. By looking at the individual differences we observed that in signal segments without artifacts, impedance can be even higher than $100k\Omega$ or lower than $10k\Omega$ for some of the participants. This indicates that more in-depth analysis is required to understand the inter-person differences in impedance, as well as how changes in impedance due to artifacts differ in case of different absolute impedance value is achieved.

The table also illustrates that the lowest changes in the impedance are introduced during head tilting and stand up/sit down movements. This was expected due to the relatively slow movement in these cases. However, contrary to our expectations, head movement in sagittal plane introduced more pronounced changes. We hypothesize that the larger force changes cause this effect, but to test this hypothesis further investigation is required. As expected, largest changes are introduced by jumping movement.

Looking at the properties of EEG signal we can see the stable mean value across sessions and participants. The impact force, walking, and jumping introduced the largest variations in the signal. Nevertheless, an important observation is that the variations in the walking and jumping sessions are three times larger than in the impact force case, and more than three times compared to the other cases. This suggests that walking and jumping conditions might be very difficult to handle as motion artifacts. Another interesting observation that needs further exploration is the fact that variations in

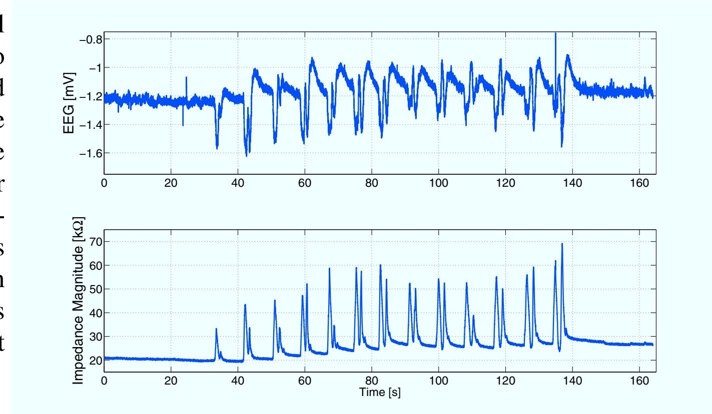

Fig. 2. EEG and impedance values recorded for Participant 6 during the coronal head movement session.

EEG during sagittal head movement are low although the impedance magnitude variations are considerably high.

Figure 3 displays the outcome of correlation analysis on the segments extracted for all sessions and for each participant. The highest correlation can be observed when applying impact force on the dry electrode. This might indicate that impedance magnitude (and EEG) can quite well describe the short-term large amplitude changes due to external force. This is further confirmed by relatively high overall correlation between the EEG and impedance in case of head movements, stand up/sit down movements, force application and release in case of continuous force application, and particularly in the correlation values for jumping segments. All these movements or forces are relatively short in time (up to 3s).

While the outcome of correlation analysis for jumping segments gives promising results, the correlation values for walking segments give less confidence that the impedance magnitude can be used for predicting such motion artifacts, with correlation values below 50% for a number of subjects. Also, for repetitive force segments lower correlation can be observed. We believe that this is due to the different dynamics reflected in EEG and impedance magnitude signal with respect to how changes in the electrode-skin interface are reflected in them. These segments are relatively long (6s or more). Also, and due to manual application of force, the absolute force could not be kept at the constant value, which in turn induced (different) variations in EEG and impedance magnitude signal. Further investigation into this effect would require more in-depth analysis and would benefit from monitoring force along with EEG and impedance magnitude.

Finally, we can observe that the correlation can be both positive and negative in each of the sessions. Typically, correlation sign is consistent per participant, but this is not always the case. Based on our preliminary analysis we were not able to infer the reason for such an effect. We stress here that before applying impedance magnitude for motion artifact reduction the reason for such inconsistency has to be revealed. This phenomena, as well as the low correlation values for some of the participants, such as for Participant 7

TABLE I

IMPEDANCE MAGNITUDE AND EEG SIGNAL PROPERTIES ACROSS EVALUATION SESSIONS. MEAN AND STANDARD DEVIATION ARE REPORTED.

		Continuous	Repetitive	Impact	Sagittal	Coronal	Standing	Walking	Jumping
Impedance magnitude $[k\Omega]$	mean	39 ± 21	35 ± 12	41 ± 20	57 ± 23	53 ± 25	52 ± 25	56 ± 27	59 ± 30
	max-min	32 ± 22	23 ± 24	35 ± 21	36 ± 33	28 ± 42	14 ± 12	36 ± 27	93 ± 261
	stdev	8.7 ± 6.5	5.1 ± 3.5	5.9 ± 5.6	7.5 ± 6.9	3.9 ± 5.2	1.5 ± 1.2	5.3 ± 5.2	7.4 ± 2.5
EEG signal $[mV]$	mean	-1.7 ± 0.17	-1.7 ± 0.09	-1.7 ± 0.20	-1.7 ± 0.22	-1.7 ± 0.23	-1.7 ± 0.28	-1.6 ± 0.29	-1.4 ± 0.61
	max-min	2.2 ± 1.34	2.6 ± 3.68	3.5 ± 1.85	1.4 ± 0.98	2.7 ± 2.58	2.0 ± 1.90	9.0 ± 4.17	9.9 ± 3.39
	stdev	0.1 ± 0.06	0.1 ± 0.11	0.2 ± 0.16	0.1 ± 0.11	0.2 ± 0.11	0.1 ± 0.13	1.0 ± 0.66	1.0 ± 0.85

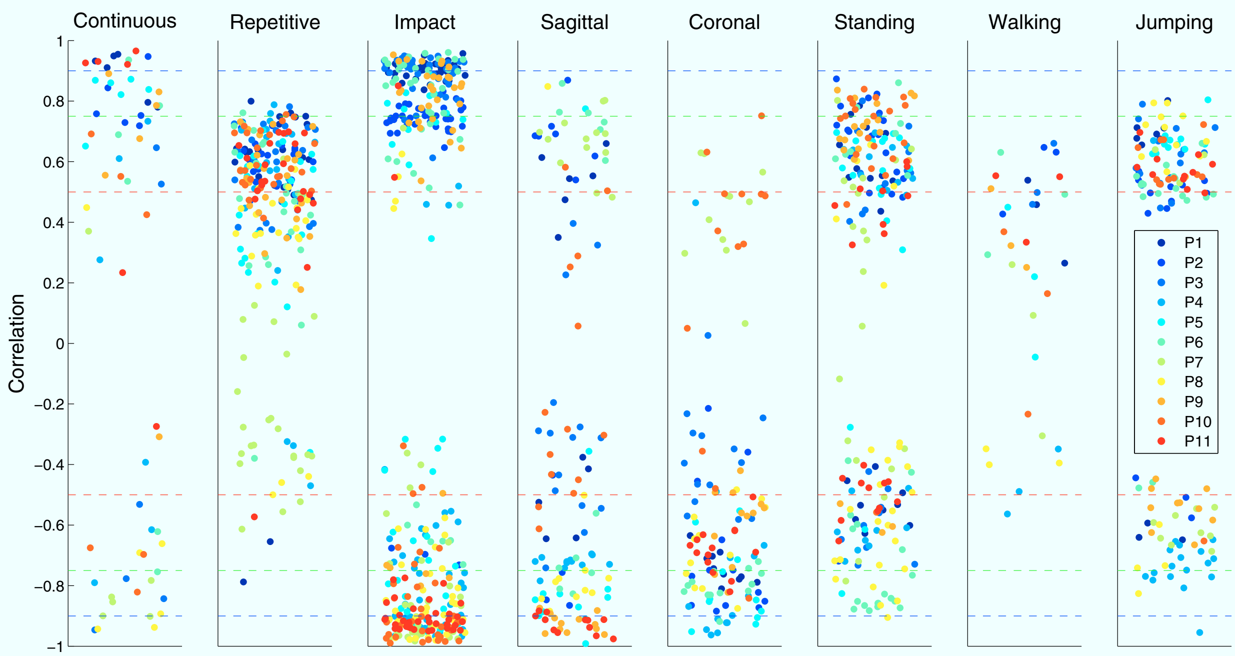

Fig. 3. Correlation between EEG and impedance magnitude across different sessions and for all participants (P1 - P11).

in case of repetitive force application and stand up/sit down movements (see Figure 3), are the ones that will be the subject of further investigations.

IV. CONCLUSIONS

The analysis shows that the correlation between EEG and impedance magnitude is high in case of short-term artifacts, especially the ones that are relatively regular such as impact force or fast movements. To a lesser degree this is true for the longer artifacts that involve smooth movements, such as head motion, as well as for the short artifacts that involve complex movement patterns, such as jumping. This demonstrates that impedance magnitude monitoring can be used for predicting EEG motion artifacts when dry electrodes are used. However, lower correlation in case of less regular artifacts demonstrates that applying continuous impedance monitoring to artifact handling in EEG would require better understanding of how these two signals, and potentially others, reflect the changes on the dry electrode-skin contact interface.

ACKNOWLEDGMENT

We thank Shrishail Patki for his assistance in constructing the evaluation setup and Ruud Vullers for his help in improving the quality of the paper.

REFERENCES

[1] Y. M. Chi, T.-P. Jung, and G. Cauwenberghs, Dry-Contact and Noncontact Biopotential Electrodes: Methodological Review, in IEEE Reviews in Biomedical Engineering, vol. 3, 2010, pp.106–119.

[2] G. Gargiulo, P. Bifulco, R. A. Calvo, M. Cesarelli, C. Jin, and A. van Schaik, A Mobile EEG System With Dry Electrodes, IEEE Biomedical Circuits and Systems Conference (BioCAS), 2008, pp. 273–276.

[3] V. Mihajlović, G. Garcia-Molina, J. Peuscher, To What Extent Can Dry and Water-Based EEG Electrodes Replace Conductive Gel Ones?, BIODEVICES, 2012, pp. 14–26.

[4] A. Nonclercq and P. Mathys, Quantification of Motion Artifact Rejection Due to Active Electrodes and Driven-Right-Leg Circuit in Spike Detection Algorithms, in IEEE Transactions on Biomedical Engineering, vol. 57, no. 11, 2010, pp. 2746–2752.

[5] J. T. Gwin, K. Gramann, S. Makeig, and D. P. Ferris, Removal of Movement Artifact From High-Density EEG Recorded During Walking and Running, in Journal of Neurophysiology, vol. 6, no. 3, 2010, pp. 3526–3534.

[6] S. Kim, H. Kim, N. Van Helleputte, C. Van Hoof, and R. F. Yazicioglu, Real Time Digitally Assisted Analog Motion Artifact Reduction in Ambulatory ECG Monitoring System, Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC), 2012, pp. 2096–2099.

[7] S. Patki, B. Grundlehner, A. Verwegen, S. Mitra, J. Xu, A. Matsumoto, J. Penders, R. F. Yazicioglu, Wireless EEG System with Real Time Impedance Monitoring and Active Electrodes, IEEE Biomedical Circuits and Systems Conference (BioCAS), 2012.

[8] V. Mihajlović and B. Grundlehner, The Effect of Force and Electrode Material on Electrode-to-Skin Impedance, IEEE Biomedical Circuits and Systems Conference (BioCAS), 2012.