Noninvasive Wearable Brain Sensing

Vojkan Mihajlović, Shrishail Patki, and Jiawei Xu
Wearable Health Solutions
Holst Centre / imec
Eindhoven, The Netherlands
{vojkan.mihajlovic, shrishail.patki, jiawei.xu}@imec-nl.nl

***Abstract*—Transforming the field of portable noninvasive brain monitoring from cumbersome, inconvenient and obtrusive sensing systems into new ergonomic, user-friendly solutions requires solid evidence from practice. Multi-modal sensing of brain signals digitized at the sensor level, minimization of noise using active sensors and shielding, and flexible and reconfigurable dry electrodes are essential components of such solutions. Usefulness of these basic ingredients has been demonstrated in several research publications, however deficiencies of novel brain sensing systems and their components over traditional gel-based systems and their clinical relevance have rarely been discussed. Here we present state-of-the-art and illustrate some of the latest developments in imec's wearable brain sensing systems and identify latest research trends and needs.**

Keywords—brain sensing; electroencephalography; EEG; wearables; noninvasive; dry electrodes; active sensors; functional near-infrared spectroscopy; fNIRS; sensors; sensor systems

I. Introduction

Noninvasive and portable brain monitoring solution exist for half a century. The first units were designed for assessing conditions of epilepsy patients by performing ambulatory electroencephalography (EEG) monitoring [1]. Since then not much has changed in the systems used to trace user's brain activity. Electrodes with conductive gel are applied by a technician at the prescribed location at the clinic and the signal quality is verified. Some form of a head cap is provided to keep the electrodes in place. Long wires were carrying the signals from a user head to the main electronic unit that user had to wear around his/her waist. All the electronics was integrated in this unit; hence it was amplifying the signal and recording it on the most suitable recording medium at the time.

Although recording units got smaller over time, it was only recently that truly wearable brain monitoring systems appear on the market, e.g., g.Nautilus (g.Tec), Smarting (mBrainTrain), and Enobio (Neuroelectrics). Advances in miniaturization of electronic components and chips was the main driver towards this change. However, cumbersome and inconvenient setup and usage of gel electrode EEG, considered as gold standard in clinical practice, is still preventing wider use of wearable brain monitoring solutions. We witness the latest revolution in wearable brain monitoring that could change the complete landscape of utilizing brain monitoring solution for numerous lifestyle and clinical applications – the use of dry electrodes. Although dry electrode sensing comes with many drawbacks in terms of increased susceptibility to noise and fragility of the interface, it comes with many potential benefits such as increased user comfort and no need for expert assistance. As such, it enables use of brain monitoring solutions at one's own pace and outside controlled clinical and laboratory environments. In this paper, we discuss the state-of-the-art in wearable dry electrode EEG sensing. We introduce flexible dry EEG electrodes and EEG sensors, as well as systems that can capture complementary brain activity signals to EEG, namely functional near infrared spectroscopy (fNIRS). We focus on identifying recent developments and future trends.

II. Dry EEG Electrodes

A. Different types of dry electrodes

To enable daily life applications and avoid injury risks, several recent wearable EEG monitoring solutions have replaced metal electrodes having a number of pins with conductive polymer electrodes with pins made by mixing carbon content into polymers [2, 3]. Typically, tips of the pins on those electrodes are coated with Ag/AgCl to improve contact properties, as discussed in next section. These electrodes (pins) are more flexible than metal counterparts, however, in case higher carbon content is used they can be quite hard. Given that higher carbon content is required for reducing impedance, there is an intrinsic flexibility/conductivity tradeoff in polymer electrodes. The performance of these electrodes in terms of contact impedance, noise and stability is still worse than conductive gel based ones but it comes close to dry rigid Ag/AgCl electrodes. This is illustrated in Fig 1 that shows the correlation and coherence in 1-40Hz frequency range of two types of coated polymer electrodes and an uncoated one to the gel electrodes, mounted 2cm apart from each other. Although Ag/AgCl coated conductive polymer electrodes are in use in commercially available system (e.g., Cognionics), their usability is not extensively evaluated. Apart from the benefits they bring in terms of user comfort and ease of integration they come with few drawbacks. In case large carbon content is used, electrodes are unlikely to follow the curvature of the head (uncomfortable) or in case low carbon content is used, the pins are completely squashed over the scalp. Furthermore, the coating on the tips of the pins can be damaged or completely removed due to the usage. Finally, the processes at the skin-electrode interface, both chemical and electrical, are not completely characterized.

B. Dry electrode to skin interface

In the absence of electrolyte, the transition of ionic tissue currents to electrode electron currents in dry contact electrodes is more complex. It results in more dominant capacitive components and overall increase of impedance to often more than 100kΩ (at 10Hz). The traditional gel electrode to skin electrical model cannot accurately describe dry electrode-tissue

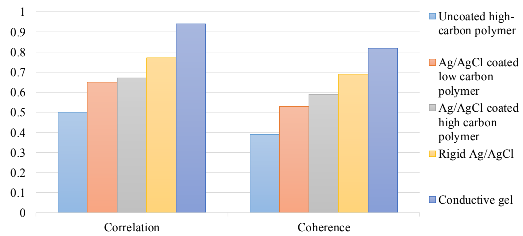

Fig. 1. Signal quality comparison between polymer, metal, and gel electrodes.

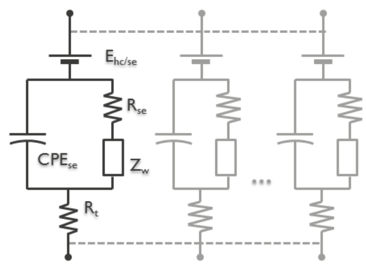

Fig. 2. Equivalent electrical circuits for dry electrode-skin contact interface.

impedance (ETI). The most suitable model for dry ETI is a result of limited analysis performed internally and it includes Warburg element (Z_w) that models the ion diffusion at the interface and constant phase element (CPE_{se}) that models the double layer at the contact interface, as illustrated in Fig. 2. This model has only been studied and demonstrated as accurate in static condition after the interface stabilization process took place. In the first minute(s) while the interface is not at equilibrium, changes are taking place. These can be seen in both, the impedance measured (see Fig. 3) as well as the low frequency drifts in the EEG. Furthermore, applying forces on the electrode introduces additional variations at the interface as electrodes can glide over the skin surface and produce various deformations of the skin beneath the electrodes. Those disturb the interface equilibrium state and are difficult to describe.

When using electrodes with pins, each pin forms a separate interface to the skin which can have different contact properties. Hence a model having a set of parallel interfaces shown in Fig. 2 would be a more realistic interpretation of the ETI model. Assisted by large differences in skin properties of general population, due to skin moisture, thickness of different skin layers, elasticity of the skin, etc., a unique model that would capture dry electrode skin interface in both static and dynamic conditions seems not feasible. Currently, we are lacking fundamental studies trying to understand the interface, especially the ones looking at conductive polymer electrodes.

C. Sensor types and layout

Dry electrodes for EEG monitoring have been mainly designed as electrodes with pins. However, in areas not covered with hair, flat coated conductive polymer electrodes would be a better solution in terms of user comfort, contact quality, and system design. Limited exploration has been done on usage of such electrodes, however, our first exploration of combining those electrodes with the pin shaped ones are promising. Also, a different sensing modalities can be supported, such as a setup that captures Laplacian derivative instead of referential signal by using concentric ring electrodes [4]. It offers a more scalable solution that has been shown also to be advantageous in terms

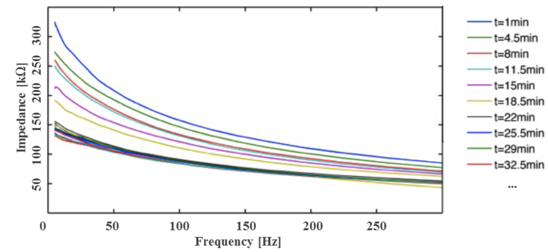

Fig. 3. Evolution of dry electrode to skin contact impedance over time.

of noise suppression for brain activity monitoring and better source localization. Also, this solution can be realized with laying out pins in the concentric ring form.

D. User comfort and risks

Introducing spring mechanisms in dry electrode EEG systems as well as introducing flexible polymer electrodes was aimed at improving user comfort over rigid metal pins. In our internal evaluation on 10 users on a scale of 1 (uncomfortable) to 10 (comfortable), average comfort of polymer electrodes was 5.83 compared to 4.17 for rigid ones. The question remains whether the comfort level is acceptable for an average consumer over a prolonged periods of time. None of the existing dry electrode solutions sustained long-term use evaluations. In our internal experiments, having users wearing latest imec's wireless EEG headset with flexible polymer electrodes, spring electrode support mechanism, and best suited headset size, complaints are often heard about the discomfort after wearing the headset for an hour. Hence, user comfort is an important parameter to assess besides electrode-skin contact properties, signal quality and stability over time, and this must be done with electrodes integrated within the adequate headset system.

Furthermore, many existing electrode and wearable headset solutions have increased risks for users. Though, preventing excessive current leakages can easily be realized, risk of mechanical injuries exists, due to rigid metal pins, rigid electrode support structure or rigid headset constructions, still exist. Also, new materials used in production of electrodes is often not tested for toxicity and sensitization on the skin. This is particularly the case with new polymer electrodes (containing carbon material) and after applying coating layers on the tips of the pins that can get in touch with damaged skin/tissue. Risk analysis and risk minimization steps are required before deploying dry electrodes in consumer products and especially in clinical environments.

III. EEG Sensors and Brain Monitoring Systems

A. Active sensors

To minimize the impact of environmental noise in already noise sensitive dry electrode EEG signal, the first signal amplification is already done immediately after the electrode-skin contact. This concept is known as active sensing and assumes that a preamplifier is positioned on a printed circuit board (PCB) cointegrated with the electrode itself (with no cables in between). Having high input impedance (typically larger than 1GΩ) and low output impedance, this active sensor ensures that the high input impedances of dry electrodes does not attenuate the EEG signal and that the artifacts due to cable motion and power line interference are minimized. This is a clear advantage over conventional solutions where artifact and interference are propagated to the signal path through the high impedance cabling connecting electrodes to the main amplifier. Although active sensors improve signal quality, they require more complex design of the sensor unit and system itself. Active sensors need to be powered and controlled, resulting in more than one wire connecting them to the backend unit. Furthermore, they require a PCB mounted on top of the electrode, increasing the overall size of the sensor unit. Hence, miniaturization and wire optimization are an integral part of active sensor design [5].

B. Noise prevention

Active sensors only partially prevent noise and artifacts that impact the EEG signal. Additional noise suppression techniques must be implemented in EEG systems. Power line interference is reduced by using driven right leg (DRL) circuitry [6]. To limit the impact of other environmental noise, special shielding mechanism are employed by using active shielding on cables. Given that most wearable brain monitoring systems use Bluetooth (BT) for data transmission, electronic components are typically protected from BT radio by proper design of the PCB layout to minimize electromagnetic interferences. These solutions prevent environment noise, but still artifacts due to movements are the main cause of noisy signals in uncontrolled recordings. Continuously monitoring contact properties in terms of impedance has been shown useful for assessing the quality of sensor contact and reducing movement artifacts [5]. Alternatively, capturing electrode or system movement using accelerometers and gyroscopes can further assist in this process. To what degree can these auxiliary signals help in extracting the signal of interest, especially when using dry electrodes, is currently an active research topic.

C. Adding optical sensing modality

Combining optical sensing (i.e., fNIRS) with EEG enables the measurement of both electrical and haemodynamic activities. Portable fNIRS systems such as Hitachi WOT-100 and Artinis Brite23, have been developed in recent years. High power consumption of these systems limits their use as fully wearable and low cost devices. fNIRS dominates the total power consumption of integrated EEG/fNIRS system, as most is dissipated by LED drivers and photodiodes to achieve sufficient optical sensing sensitivity [7]. One promising solution to mitigate the tradeoff between power and sensitivity of a fNIRS system is the new generation photodiode, known as silicon photomultiplier (SiPM), Although SiPM requires a high bias voltage (~30V), LED driving current is substantially reduced from 10-50mA to 10-100µA. The latest imec active electrode ASIC that uses low power multimodal sensing channels facilitates simultaneous and active recording of both EEG and fNIRS. The ASIC can be mounted with an electrode/optode as a multimodal active sensor.

D. Signal digitization

For multi-channel recordings, interface between active sensors and the backend readout must be simple and flexible, such that the number of connecting wires is minimized and that the interface can easily be scaled to support a larger number of channels. Conventional active sensors with analog outputs require many wires connected to the backend unit and are limited by the number of channels the readout can process. This results in a complex system when multimodal and multichannel recording is required. Latest imec wearable brain monitoring systems utilize digital active electrode (DAE) concept [8]. Each electrode is co-integrated with a low power ASIC equipped with a built-in ADC and an I^2C interface. Thus, the EEG signal is pre-amplified and digitized locally before transmitting it to the backend unit. Multiple DAEs can be connected via an I^2C bus to a generic microcontroller in a daisy chain or a star architecture. This solution reduces both system complexity and cost, as custom designed analog backend readout is not required. The DAE and its I^2C communication provides flexibility to add or remove electrodes based on the application. Additionally, the system architecture with DAE ensures that no changes are necessary to interface with future generation DAEs with additional functionality such as new optical or current/voltage sensors, making the architecture highly modular. Furthermore, signal quality is improved because the digital outputs are inherently more robust to interferences than analog ones.

E. Signal processing

Besides simple filtering and resampling, other methods to process acquired signals are rarely implemented within wearable EEG or fNIRS systems. The data analysis is mainly done on a PC, having more operating power, dedicated tools, and flexibility to test and visualize output of different analysis methods. Having more powerful microprocessors, such as Cortex M4F in latest imec's wireless EEG headsets, facilitates implementation of more complex signal processing algorithms within the system. This can include not only the ones for extracting spectral power or independent components but also advanced algorithms for reducing the impact of noise and artifacts in runtime. Only a few examples exist that perform embedded information extraction, new dedicated applications and wider adoption of wearable brain monitoring raise the need for efficient embedded algorithm implementation.

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