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Noninvasive Wearable Brain Sensing

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***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 and prospects of novel brain sensing systems and their components over traditional gel-based systems and their clinical relevance have rarely been discussed. Here, we present the 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. Special emphasis is given to recent technological developments that enable multi-modal wearable brain sensing by combining EEG with functional near infrared spectroscopy and electrical impedance tomography.**

Keywords—brain sensing, electroencephalography, EEG, wearables, noninvasive, dry electrodes, active sensors, functional near-infrared spectroscopy, fNIRS.

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 locations at the clinic and the signal quality is verified. 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 appeared on the market, e.g., g.Nautilus (g.Tec) [2], Smarting (mBrainTrain) [3], and Enobio (Neuroelectrics) [4]. Advances in miniaturization of electronic components and chips were the main driver towards this change. However, cumbersome and inconvenient setup and usage of gel electrode, considered as gold standard in clinical practice for EEG acquisition, are still preventing wider use of wearable brain monitoring solutions. We witness the latest revolution in wearable brain monitoring that could change the landscape of utilizing brain monitoring solution for 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, short setup times, and no need for expert assistance. As such, it enables the use of brain monitoring solutions at one's own pace and outside controlled clinical and laboratory environments.

The first systems that used dry electrodes for monitoring electrical activity of the brain appeared a decade ago [5]-[7]. Those were research prototypes, demonstrating the potential of dry EEG monitoring. It was not long after that the first commercial devices appeared on the market, such as NeuroSky [8] and Quasar [9]. Those devices paved the way for many new dry electrode solutions appearing recently at the market, mainly used to capture the status of human brain in daily life situations or when humans are put in specific environments [10][11].

EEG has limitations in terms of spatial resolution, especially if a limited number of electrodes is used, and capturing activity of deeper brain regions is a must. To complement monitoring of the electrical activity of the brain, blood oxygenation monitoring through functional near infrared spectroscopy (fNIRS) has shown to provide clinically relevant information [12]. Given the potential of miniaturization of fNIRS systems, several wearable integrated EEG and fNIRS systems for monitoring human brain activity in a convenient way have recently appeared [13][14]. One of the most recent developments introduces also the possibility to capture bioimpedance information, which enables electrical impedance tomography (EIT) monitoring [15].

Combining electrical and optical sensing has the potential to improve the performance of EEG based brain-computer interface (BCI) systems in terms of detection accuracy [16]. Also, it facilitates reliable prediction of drivers' fatigue level [17]. Recently, the benefits of simultaneous EEG and fNIRS monitoring in clinical applications have been demonstrated on assessing neurovascular coupling in infants [18]. Further examples of use cases are missing mainly due to the unavailability of such multi-modal monitoring solutions, especially the ones including EIT. It is a matter of time before we will witness more success stories utilizing such convenient, multi-modal, wearable brain monitoring technology.

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 and discuss the main aspects of wearable dry EEG sensing technology. We also provide more details on the technological solutions for multi-modal brain activity monitoring. In addition, we focus on identifying recent developments and future trends.

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II. Dry EEG Electrodes

A. *Different types of dry electrodes*

The first dry (contact) electrodes were rigid, made of stainless steel, silver/silver-chloride (Ag/AgCl), or gold-plated material [19]. Flat electrodes were used for head regions which are not covered with hair, while a rigid metal pin structure was used for hairy sites. Among those, Ag/AgCl electrodes have shown the best performance and are used in most solutions [19]. As a certain amount of flexibility and comfort are required, spring-like mechanisms were introduced, either within the pins or on top of the electrodes.

Recently, to enable daily life applications and avoid injury risks, several wearable EEG monitoring solutions have replaced metal electrodes with conductive polymer electrodes containing pins, made by mixing carbon content into polymers [20][21]. Typically, tips of the pins on those electrodes are coated with Ag/AgCl to improve contact properties, as discussed in the next section. These electrodes (pins) are more flexible than their metal counterparts, however, when higher carbon content is used they can be quite rigid. Given that higher carbon content is required for reducing contact impedance, there is an intrinsic flexibility/conductivity tradeoff in polymer electrodes. The performance of these electrodes is still worse than conductive gel based ones in terms of contact impedance, noise and stability, but it comes close to dry rigid Ag/AgCl electrodes. This is illustrated in Fig. 1 that shows the correlation and coherence in 1-40 Hz frequency range of two types of coated polymer electrodes and an uncoated one to the gel electrodes, mounted 2 cm apart from each other.

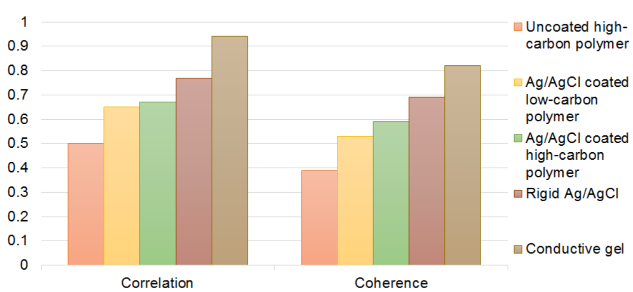

Fig. 1. Correlation and coherence of EEG signals measured using a reference conductive gel electrode and different test electrodes with pins placed 2cm apart: uncoated polymer, Ag/AgCl coated polymer (with low and high carbon content), rigid Ag/AgCl electrode and conductive gel electrode.

Although Ag/AgCl coated conductive polymer electrodes are in use in commercially available system (e.g., Cognionics [22]), their usability is not extensively evaluated. Apart from the benefits these conductive polymer electrodes bring in terms of user comfort and ease of integration, they come with a few drawbacks. In case large carbon content is used, electrodes are unlikely to follow the curvature of the head resulting in poor electrode-skin contact and user discomfort. In case low carbon content is used, penetrating the hair and reaching the scalp becomes difficult. 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 is more complex in dry electrodes. It results in more dominant capacitive components and overall increase of impedance to often more than 100 kΩ (at 10 Hz) [23]. The traditional gel electrode to skin electrical model shown in Fig. 2a cannot accurately describe dry electrode-tissue impedance (ETI). Most importantly, the electrode-electrolyte (E_{hc}, C_{ee} and R_{ee}) and electrolyte-skin interface (E_{se}, C_{se} and R_{se}) does not exist.

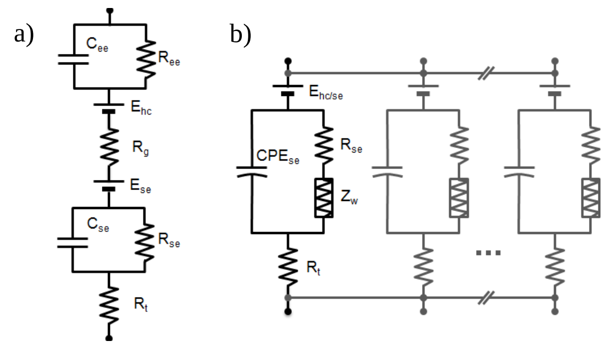

Fig. 2. Equivalent electrical circuits for: a) conductive gel electrode-skin interface and b) dry electrode-skin contact interface. E_{hc} is half cell potential, C_{ee} and R_{ee} are electrode-electrolyte capacitance and resistance, C_{se} and R_{se} are electrolyte-skin interface, CPE_{se} is dry electrode-skin constant phase element (capacitive component), Z_w is dry electrode-skin Wartburg element, and R_t is skin tissue resistance.

The most suitable model derived for dry ETI is a result of a limited analysis performed at imec. It includes a 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. 2b. This model has only been studied and demonstrated as accurate in static conditions and when using metal electrodes with rigid metal pins on the surface of the skin.

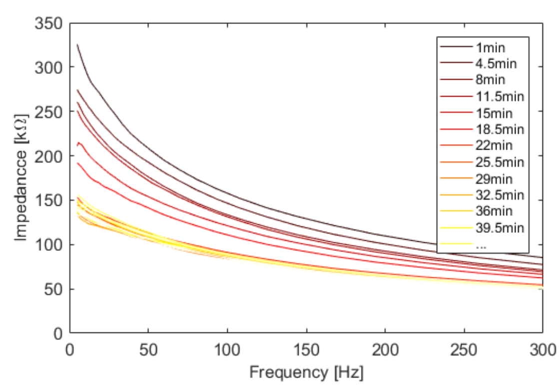

Fig. 3. Evolution of dry electrode to skin contact impedance over time. Each line represents impedance magnitude measured in the 1-300 Hz frequency range at a different time instance starting from the moment measurement electrodes are mounted on the head.

Furthermore, these measurements were done after the interface stabilization process took place. In the first minute(s), while the interface is not at equilibrium, changes are taking place at this interface. These can be seen when inspecting the evolution of the impedance over time, as shown in Fig. 3. In case of such dry electrodes, it takes more than 20 minutes for the impedance values to stabilize. In addition to the long impedance stabilization time, large drifts at low frequency are observed in the first few minutes when EEG measurements are performed with dry electrodes. Furthermore, applying different 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 [24]. Those forces disturb the interface equilibrium state and are difficult to describe and model.

When using electrodes with pins, each pin forms a separate interface to the skin which can have different contact properties. This assumes a different force is applied on each pin, as well as different contact area is formed, including the potential impact of hair. Hence, a model having a set of parallel interfaces shown in Fig. 2b would be a more realistic interpretation of the ETI model. So far, only limited exploration has been done on polymer electrodes, focusing mainly at treating the interface of a number of pins as a single interface [21]-[25]. Although, the results of such studies are useful for the application and selection of the best electrode configuration, they are not helpful in understanding the interface under static and dynamic conditions. Complemented by large differences in skin properties of the general population, due to skin moisture, thickness of different skin layers, elasticity of the skin, etc., generating a suitable unified model that would capture dry electrode skin interface in both static and dynamic conditions remains a big challenge.

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 the usage of such electrodes. In our initial evaluation, the typical impedance level of such electrodes was similar to the rigid Ag/AgCl electrodes with pins. The mean values across the ten participants using the EEG system with different dry electrodes integrated in a headset described in [26] during the 30-minute recording was: 6.5 kΩ for reference gel electrodes, 50.5 kΩ for flat Ag/AgCl coated conductive polymer electrodes, 58 kΩ for rigid Ag/AgCl electrodes with pins and 62.2 kΩ for Ag/AgCl coated conductive polymer electrodes with pins.

Apart from flat and single multi-pin electrodes, different electrode embodiments can be implemented using such flexible flat or pin electrode structures. Such a setup can be used to capture the Laplacian derivative instead of referential signal by using concentric ring electrodes [27]. Conductive gel based concentric ring electrodes offer a more scalable solution than the conventional gel ones. Also, they are advantageous in terms of noise suppression for brain activity monitoring and provide better source localization [28]. Dry flat concentric rings are an alternative option to those. A Laplacian solution can also be realized by laying out the pins in the form of a concentric ring. However, these research directions are not yet fully explored.

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, average comfort of polymer electrodes was 5.83 compared to 4.17 for rigid ones with springs, rated on a scale of 1 (uncomfortable) to 10 (comfortable). The results on the user comfort while wearing an EEG system described in [26], where different dry electrodes are introduced, are given in Fig. 4. They indicate higher comfort for both flat and pin flexible polymer electrodes, rigid metal electrodes. The question remains whether the comfort level is acceptable for an average consumer over prolonged periods of time. None of the existing dry electrode solutions sustained long-term use evaluations. In our internal experiments, having users wearing the latest imec's wireless EEG headset with flexible polymer electrodes, spring electrode support mechanism, and best suited headset size, complaints about discomfort are often raised after wearing the headset for about an hour. Hence, user comfort is an important parameter to assess, besides electrode-skin contact properties, signal quality and stability over time. This evaluation must be done, considering also electrodes integrated within an adequate headset system, as the headset properties can have substantial impact on the comfort.

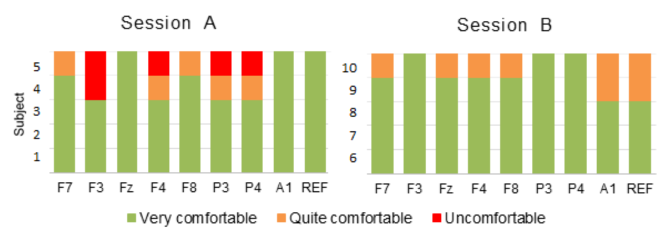

Fig. 4. Usser rating of comfort level during two measurement sessions (10 users) of around 30 minutes and across 9 electrode locations. F3 and P3 positions in Session A have rigid Ag/AgCl electrodes and the rest are different version of conductive polymer electrodes with Ag/AgCl coating, either flat on electrode positions F7 or F7 or with pins (at the rest of the positions).

Furthermore, many existing electrode and wearable headset solutions have increased risks for users. While excessive current leakage can easily be prevented, a risk of mechanical injuries still exists, due to the rigid metal pins, rigid electrode support structure or rigid headset constructions. Also, new materials used in the production of electrodes are 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 may get in contact with damaged skin or tissue. Risk analysis and risk minimization steps are required before deploying dry electrodes in consumer products and especially before any medical use. As a minimum requirement before the use of any of the newly developed electrodes, they need to be proven to be non-toxic when used on humans. Toxicity and risk analysis are regular procedures followed in imec before applying new electrodes and headsets on people.

III. EEG Sensors and Brain Monitoring Systems

A. Active sensors

To minimize the impact of environmental noise on the noise sensitive dry electrode EEG signal, the first signal amplification is 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 minimal wire routing in between). Due to high input impedance and low output impedance, this active sensor ensures that the high impedance 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 passive 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 the 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 wiring optimization are an integral part of active sensor system design [29].

B. *Noise prevention*

Active sensors only partially prevent noise and artifacts that impact the EEG signal. Due to low signal amplitudes, additional noise suppression techniques must be implemented in EEG systems. Environmental common mode interference can be reduced by using the driven right leg (DRL) circuitry [30]. Precautions are also taken to protect the sensitive analog traces. They are protected from relatively high-speed clock signals by a careful layout, ensuring minimal interference from switching clock signals. To limit the impact of other environmental noise, active shielding [31] mechanisms are used on the cables.

Given that most wearable brain monitoring systems use Bluetooth (BT) for data transmission (e.g, Cognionics, Enobio, NeuroSky), electronic components are typically protected from the radio by a careful PCB layout to minimize electromagnetic interferences. Due to system miniaturization in wearables, it is a challenge to position the radio away from the sensitive analog circuits. The high frequency radio signal at 2.4 GHz may appear in the bandwidth of EEG (up to 100 Hz) due to aliasing as electro-magnetic interference (EMI). The interference can be as large as few µVs. Considering that the EEG amplitude is also at a similar level, the noise becomes significant if not filtered properly. To address the EMI and improve the RF immunity at acquisition electrodes, low pass filters with a cut-off frequency around 1 MHz are often implemented very close to the electrodes, in order to filter the high frequency noise at the source itself. Additionally, it is necessary to improve the RF immunity at the bias or patient ground electrode, hence a low pass filter is also implemented at the bias electrode. This is illustrated in Fig. 5. The interference also depends on the output power of the radio, typically set at 0 dBm. Reducing this output power, in addition to the careful layout and proper grounding techniques, helps in reducing EMI during the analog signal acquisition.

These solutions reduce the impact of environment noise, but other noise sources can still impact the EEG signal. Among those, the strongest interference is due to the artifacts caused by movements. They produce substantial disturbances in the gel-based EEG recordings, even larger when dry electrodes in a wearable headset form factor are deployed, and in particular when the system is used in uncontrolled environments. Continuous monitoring of ETI has shown useful insights for assessing the quality of sensor contact [29] but also in compensating motion artifacts [32]. Alternatively, capturing electrode or system movement using accelerometers and gyroscopes can further assist in this process [33]. To what degree these auxiliary signals can help in extracting the signal of interest, especially when using dry electrodes, is currently an active research topic.

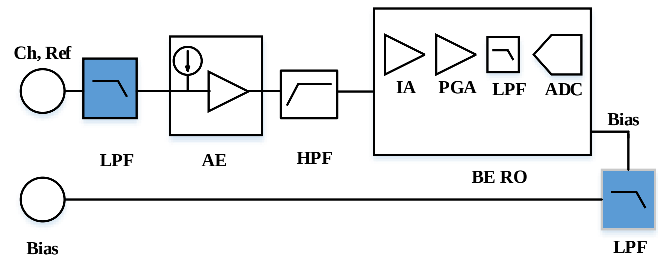

Fig. 5. High-level block diagram including EMI minimization: introduction of RC low pass filters (LPF) between channel (Ch) and reference (Ref) dry electrodes and active electrode (AE) integrated circuits (ICs), as well as between bias output from back-end readout (BE RO) IC and bias electrode. The high pass filter (HPF) shown in the diagram is used for ensuring AC coupling between AE and BE.

C. *Signal digitization*

For multi-channel EEG recordings, the interface between the 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 backend readout can process. This results in a complex system when multimodal and multichannel recordings are required. Imec's latest wearable brain monitoring systems use a digital active electrode (DAE) concept [34] that is illustrated in Fig. 6.

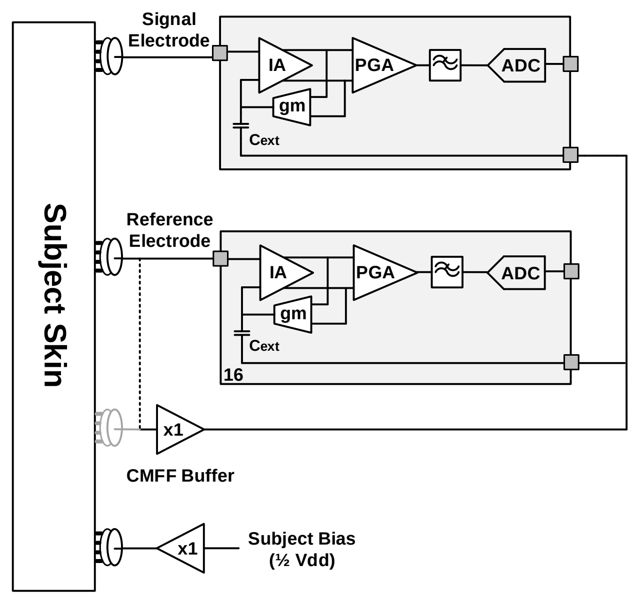

Fig. 6. Digital active electrode (DAE) concept block diagram.

Each electrode is co-integrated with a low power application-specific integrated circuit (ASIC) equipped with a built-in ADC and a digital interface. Thus, the EEG signal is pre-amplified and digitized locally before it is transmitted to the backend unit. Multiple DAEs can be connected via mixed-signal bus to a generic microcontroller in a daisy chain or a star architecture. A common mode feed forward (CMFF) signal, i.e., a buffered output of reference electrode, is used as a common mode input to all the DAEs, improving the common mode rejection ratio (CMRR) [34]. This solution reduces both system complexity and cost, since an analog signal processing back-end readout is not required.

D. Signal processing

Besides filtering and resampling, other methods to process acquired signals are rarely implemented within wearable EEG (or fNIRS) systems. The data analysis is mainly done externally 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 imec's latest wireless EEG headsets, facilitates implementation of more complex signal processing algorithms within the system. This can include not only the extraction of 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 [35]. New dedicated applications and wider adoption of wearable brain monitoring (e.g., [36]) raise the need for efficient embedded algorithm implementations.

IV. MULTIMODAL BRAIN SENSING

EEG measures the biopotential signals of a large group of neurons, having the advantage of excellent temporal resolution (<1 ms). However, EEG suffers from poor spatial resolution because the signal is measured on the scalp, and is hence blurred by different conduction layers in between the scalp and the cortex including skull, cerebrospinal fluid, etc. The spatial resolution of brain activity monitoring is improved when other noninvasive neuroimaging techniques are used, such as positron emission tomography (PET), functional magnetic resonance imaging (fMRI) and functional near infrared spectroscopy (fNIRS). These techniques are used to measure hemodynamic responses of brain tissues, e.g. the cerebral blood flow due to increased neuronal activity, with a high spatial resolution (1-2 mm). However, PET and MR scanners are bulky, require special shielded rooms, are extremely power hungry, and costly to produce and maintain. In contrast, state-of-the-art portable and wearable fNIRS systems [37][38] are by far less complex in design, have lower power consumption, and can be miniaturized. However. they cannot yet facilitate long-term and continuous fNIRS measurements. Combining both EEG and fNIRS [15][39][40] and potentially other modalities (e.g. electrical impedance tomography) provides complementary results to understand the brain activities by using a compact and low-cost solution.

A. Functional near-infrared spectroscopy (fNIRS)

fNIRS measurement is based on the modulation of near infrared light in the spectrum of 700-900 nm. Pulsed light trains from a two-colour LED is shined into the tissue (Fig. 7), where the light is modulated based on the hemodynamic or blood-oxygen-level dependent response. The light travelling through the tissue is then detected by an optical receiver and is further converted into an output pulsed current.

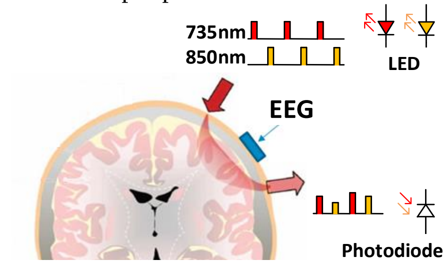

Fig. 7. Illustration of a multimodal EEG and fNIRS recording, including the LED pulse train and the light pulses captured by the photodiode.

The optical receiver can be implemented with a photodiode (PD), which has the advantage of a small size and a low bias voltage (<5V), but the limited sensitivity of a PD requires a few mW of power consumed from the LED even if it is duty cycled. An avalanche photodiode (APD) has a higher light sensitivity, but an APD requires a bias voltage between 50-100 V, which is a challenge for low power wearable devices. Recently introduced silicon-photomultiplier (SiPM), i.e., a pixelated based APD array, has shown to provide high light sensitivity [41], relaxes the power of the LED driver, and is thus a promising solution for power-efficient fNIRS measurement. Furthermore, the SiPM can detect the light pulses travelling through a deeper or a wider region of the brain tissue, allowing for a reduced number of transmitters and receivers to cover the whole scalp area. This reduces the power of the system significantly, even though the SiPM requires a bias voltage of 30-35 V. When considering the significant power saved in the LEDs (reduction from 10-50 mA to 10-100 µA), an SiPM-based optical sensing system still provides better overall power efficiency than a PD-based system [42].

The readout circuit interfaced with a PD or an SIPM can be a transimpedance amplifier (TIA) or an active integrator (INT) (Fig. 8). The output voltage is sampled by an ADC. The fNIRS representing hemodynamic responses is usually sampled below 16Hz, however, either the analog outputs or the fNIRS pulses are often oversampled/averaged for improved resolution.

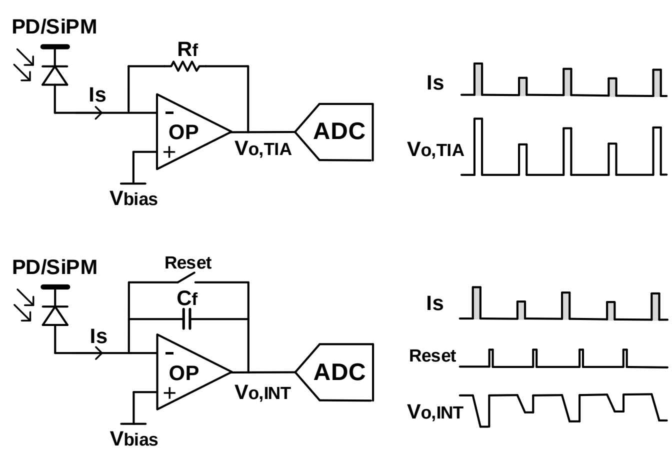

Fig. 8. fNIRS readout circuits based on a TIA (above) or an integrator (below). C_f represents feedback integration capacitor, Is is the current from PD/SiPM,

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and $V_{o,TIA}$ and $V_{o,INT}$ are the output voltage of TIA and integrator, respectively. Waveforms at the major circuit nodes are given on the right.

A major challenge for fNIRS sensing is the large dynamic range of at least 80 dB required due to the ratio between the large DC signal (i.e. ambient light and static fNIRS signals) and the small AC signal (i.e. dynamic fNIRS signals). State-of-the-art fNIRS acquisition circuits employ a (pulsed) compensation current to improve dynamic range, but the current is often set manually [40][42]. A digitally-assisted feedback loop, developed at imec [15] (Fig. 9), runs in the background and provides the pulsed compensation current automatically. This loop, operating in the digital domain, is more power and area efficient, while the full range fNIRS input signal can be reconstructed by combining the ADC outputs and the amount of compensation current.

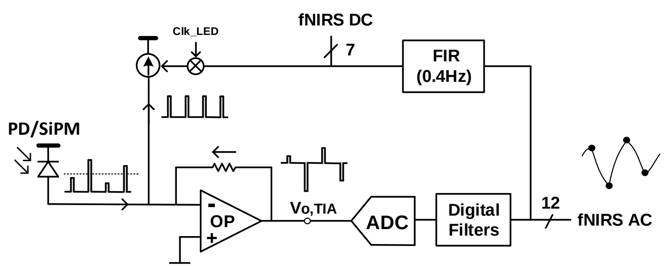

Fig. 9. Block diagram of a digitally-assisted feedback loop for improved dynamic range.

The fNIRS system described above can detect the heart rate with a single-wavelength LED by measuring the blood volume variability (Fig. 10). To further measure the hemodynamic responses, such as SpO_2 for blood oxygen level estimation, two-wavelength LEDs are required. Fig. 11 shows the relative SpO_2 change of the subject when holding breathing (from 5 s to 46 s) and restarting breathing (after 46s). The measurement was done on the forehead with a SiPM and a two-wavelength LED. The trend of SpO_2 change is in a good agreement with the subject's breathing pattern. A more comprehensive 3D brain imaging can be obtained by placing multiple optodes (LED and SiPM pairs) over the scalp.

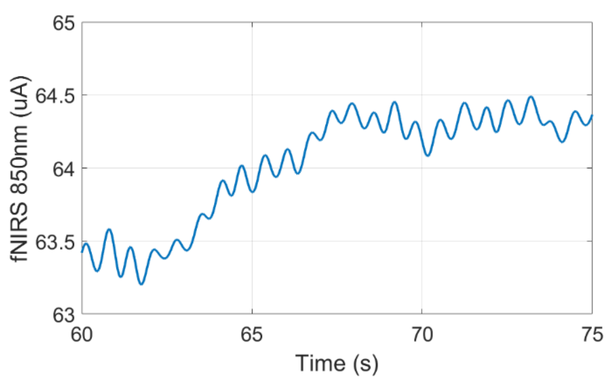

Fig. 10. The fNIRS currents measured from the subject's forehead when a single LED (850 nm) is used.

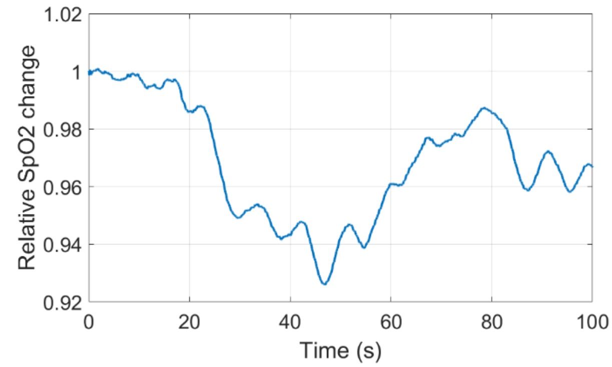

Fig. 11. Relative SpO_2 derived from a measurement with two wavelength (735 nm, 850 nm) fNIRS recording.

B. *Electrical impedance tomography (EIT)*

Compared to the fMRI and fNIRS, the EIT of the brain can provide fast impedance imaging (<1 ms) to assess the status during or after certain cognitive dysfunction, such as stroke or epileptic seizure, in a low-cost manner. A big challenge of non-invasive brain EIT is the high-impedance introduced by the skull, which attenuates the current going through intracranial tissue. However, successful EIT with scalp electrodes has been realized with advanced signal processing and modeling techniques [43].

The EIT sequentially measures the bio-impedance between multiple electrode pairs. Thus, the brain impedance network is obtained in a time-multiplexed manner. The overall frame rate of n-electrode EIT should be more than 1 kHz to detect neural activities of the brain. Bio-impedance measurement is based on current-to-voltage conversion (Fig. 12), where a pair of electrodes generates excitation current and another pair of electrodes measures the voltage across the bio-impedance. Such a 4-electrode configuration can mitigate the error introduced by the ETI. This helps to relax the dynamic range requirement of the readout circuit, and reduces noise. However, to sense the small bio-impedance change superimposed on the large skull impedance, the readout circuit needs to provide a large dynamic range (estimated to be more than 100 dB) with a high sensitivity (estimated to be less than 10 mΩ).

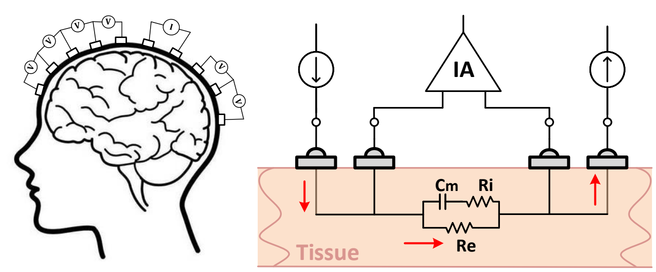

Fig. 12. Multi-electrode EIT setup on a participant's head and a 4-electrode-based bio-impedance sensing unit including the tissue model. C_m is membrane capacitance, R_i is intracellular resistance, and R_e is extracellular resistance.

Imec's solution to improve sensitivity is to reduce the system noise. As bio-impedance variation spans from sub-Hz to a few kHz, the flicker noise or 1/*f* noise of the system becomes the bottleneck for sensitivity. Conventional EIT readout circuits only focus on noise reduction of the sensing amplifier, while the

1/*f* noise of the excitation current source can be a dominant noise source. In order to solve this issue, a dynamic element matching (DEM) technique is applied to the excitation current generator to mitigate its 1/*f* noise by upmodulating it to a higher frequency (Fig. 13) [15]. Measurement tests demonstrated the benefits of such a solution, however its practical evaluation on users still needs to be explored.

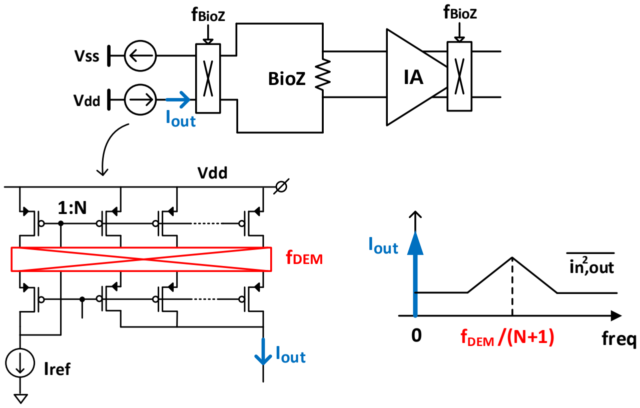

Fig. 13. Excitation current source with reduced 1/*f* noise. Block diagram of bio-impedance measurement (top), where f_{BioZ} is the measurement frequency of BioZ. Transistor level low-noise excitation current source (bottom left), where I_{out} is the output sourcing current of the CG, f_{DEM} is the frequency of dynamic element matching. Illustration of the current source noise versus frequency (bottom right).

C. Multimodal sensor integration and measurements

Fig. 14 shows a conceptual sketch of a module integrating EEG, fNIRS and EIT sensors and a readout ASIC. Each module contains two electrodes, one for EEG (and EIT) electrical potential signal sensing and the other one for EIT current sinking/sourcing. A dual-wavelength LED and SiPM (or PD) are placed on electrode sides. Both optodes have optical isolation to minimize interference and reduce the impact of ambient light. All these sensors are connected to the ASIC mounted on top of the EEG electrode.

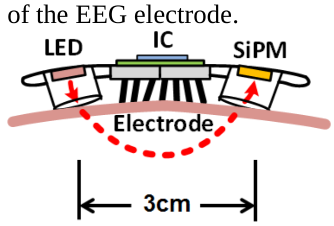

Fig. 14. Illustration of a multimodal EEG, fNIRS and EIT sensor module.

The proposed system also supports simultaneous and multi-channel EEG, fNIRS and EIT measurement (Fig. 15). Bipolar EEG or EIT requires a total of N+1 modules to realize N-channel measurement, where one module is selected as the reference. Thus, EEG or EIT is acquired between modules. Unipolar fNIRS requires N modules for N channel, as the LED and SiPM are both located on the same module.

During the multimodal measurement, EEG is continuously measured at all sensor modules, while fNIRS is measured in a time-multiplexing manner. The LED of each module is enabled sequentially, and multiple SiPMs around one LED detect the light simultaneously. This inter-module sensing increases the number of fNIRS channel, having shared optodes, leading to a compact system. EIT is also measured in a time-multiplexing manner. A pair of modules is enabled each time and they provide activation of sinking and sourcing currents, respectively. The bio-impedance signal between these two modules is measured. Similar measurement can be repeated among different module pairs for a full head EIT. Note that the EEG and EIT can share the same electrode for voltage sensing because they have different bandwidth.

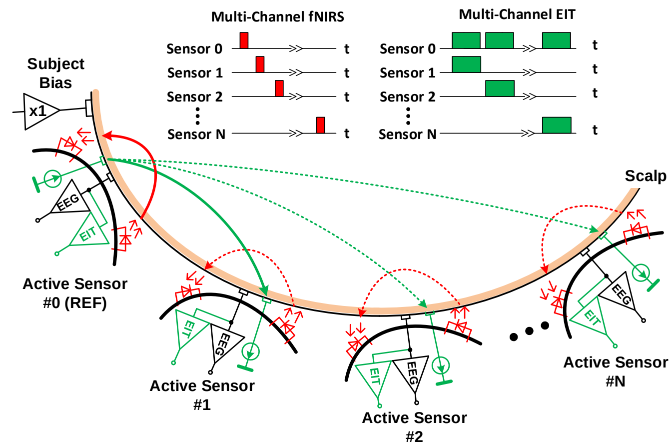

Fig. 15. Illustration of a multichannel EEG, fNIRS and EIT measurement.

V. CONCLUSIONS

The latest developments in the area of noninvasive and wearable brain monitoring are discussed in this paper. The focus is on the dry EEG electrode designs and brain monitoring system aspects required to provide reliable capturing of brain state while allowing for a greater comfort to the user. Dry electrode EEG solution challenges in terms of electrode and sensors design, noise minimization and signal digitization aspects are presented. Additionally, new directions in terms of multi-modal brain monitoring are introduced, including several system solutions on how to implement miniaturized, low power functional near-infrared spectroscopy and electrical impedance tomography. We illustrate how these three sensing modalities can be integrated in a single system and the benefits they bring. Addressing the challenges within the multi-modal brain monitoring solutions is the essential ingredient for the next generation of wearable brain monitoring devices.

REFERENCES

[1] W. O. Tatum, Ed., *Ambulatory EEG Monitoring*, New York, NY, USA: Springer, 2017.

[2] g.Tec, http://gtec.at/, accessed on May 17th, 2018.

[3] mBrainTrain, https://mbraintrain.com, accessed on May 17th, 2018.

[4] Neuroelectrics, https://www.neuroelectrics.com/, accessed on May 17th, 2018.

[5] F. Popescu, S. Fazli, Y. Badower, B. Blankertz, K.-R. Müller, “Single Trial Classification of Motor Imagination Using 6 Dry EEG Electrodes,” *PLoS ONE*, vol. 2, no. 7, e637, 2007.

[6] G. Gargiulo, P. Bifulco, R. A. Calvo, M. Cesarelli, C. Jin, A. van Schaik, “A mobile EEG system with dry electrodes”, *IEEE Biomedical Circuits and Systems Conference (BioCAS)*, 2008.

[7] E. W. Sellers, P. Turner, W. A. Sarnacki, T. McManus, T. M. Vaughan, R. Matthews, “A Novel Dry Electrode for Brain-Computer Interface,”

Human-Computer Interaction. Novel Interaction Methods and Techniques, pp. 623-631, 2009.

[8] NeuroSky, http://neurosky.com/, accessed on March 25th, 2018.

[9] Quasar, http://quasarusa.com/, accessed on March 25th, 2018.

[10] D. E. Callan, G. Durantin, C. Terzibas, “Classification of single-trial auditory events using dry-wireless EEG during real and motion simulated flight,” *Frontiers in Systems Neuroscience*, vol. 9, no. 11, 11, 2015.

[11] P. Pradhapan, U. Großekathöfer, G. Schiavonne, B. Grundlehner, V. Mihajlović, “Toward Practical BCI Solutions for Entertainment and Art Performance,” in C.S. Nam, A. Nijholt, F. Lotte, Ed., *Brain-Comupter Interfaces Handbook*, CRC Press, 2018.

[12] A. M. Kempny, L. James, K. Yelden, S. Duport, S. Farmer, E. Diane Playford, A. P. Leff, “Functional near infrared spectroscopy as a probe of brain function in people with prolonged disorders of consciousness,” *NeuroImage: Clinical*, vol. 12, pp. 312-319, 2016.

[13] A. Kassab, J. Le Lan, J. Tremblay, P. Vannasing, M. Dehbozorgi, P. Pouliot, A. Gallagher, F. Lesage, M. Sawan, D. K. Nguyen, “Multichannel wearable fNIRS ‐ EEG system for long ‐ term clinical monitoring,” *Human Brain Mapping*, vol. 39, no. 1, pp. 7-23, 2018.

[14] A. von Lühmann, H. Wabnitz, T. Sander, K.-R. Müller, “M3BA: A Mobile, Modular, Multimodal Biosignal Acquisition Architecture for Miniaturized EEG-NIRS-Based Hybrid BCI and Monitoring,” *IEEE Transactions on Bimedical Engineering*, vol. 64, no. 6, pp. 1199-1210, 2017.

[15] J. Xu et al., “A 665µW silicon photomultiplier-based NIRS/EEG/EIT monitoring ASIC for wearable functional brain imaging,” *Digest of ISSCC*, pp. 294-295, 2018.

[16] S. Ahn, S. C. Jun “Multi-Modal Integration of EEG-fNIRS for Brain-Computer Interfaces – Current Limitations and Future Directions,” *Fronteers in Human Neuroscience*, vol 11, no. 503, 2017.

[17] S. Ahn, T. Nguyen, H. Jang, J. G. Kim, S. C. Jun, “Exploring neuro-physiological correlates of drivers’ mental fatigue caused by sleep deprivation using simultaneous EEG, ECG, and fNIRS data,” *Fronteers in Human Neuroscience*, vol 10, no. 2019, 2016.

[18] E. Bourel-Ponchel, M. Mahmoudzadeh, A. Delignières, P. Berquin, F. Wallois “Non-invasive, multimodal analysis of cortical activity, blood volume and neurovascular coupling in infantile spasms using EEG-fNIRS monitoring,” *NeuroImage: Clinical*, vol 15, pp. 359-366, 2017.

[19] M. A. Lopez-Gordo, D. Sanchez-Morillo, F. Pelayo Valle, “Dry EEG Electrodes,” *Sensors*, vol. 14, no. 7, pp. 12847-12870, 2014.

[20] Y. M. Chi, Y. Wang, Y.-T. Wang., T. P. Jung, T. Kerth., Y. Cao, “A Practical Mobile Dry EEG System for Human Computer Interfaces,” *LNCS*, vol. 8027. Springer, pp. 649-655. 2013.

[21] Y.-H. Chen, M. Op de Beeck, L. Vanderheyden, E. Carrette, V. Mihajlović, K. Vanstreels, B. Grundlehner, S. Gadeyne, P. Boon and C. Van Hoof, “Soft, Comfortable Polymer Dry Electrodes for High Quality ECG and EEG Recording,” *Sensors*, vol. 14, pp. 23758-23780, 2014.

[22] Cognionics, https://www.cognionics.net/, accessed on May 17th, 2018.

[23] A.-M. Tautan, V. Mihajlović, Y.-H. Chen, B. Grundlehner, J. Penders, and W. Serdijn, “Signal Quality in Dry Electrode EEG and the Relation to Skin-electrode Contact Impedance Magnitude.” *International Joint Conference on Biomedical Engineering Systems and Technologies* (BIOSTEC), pp. 12-22, 2014.

[24] V. Mihajlović, B. Grundlehner, “The effect of force and electrode material on electrode-to-skin impedance,” *IEEE Biomedical Circuits and Systems Conference*, 2013.

[25] P. Fiedler, D. Strohmeier, A. Hunold, S. Griebel, R. Mühle, M. Schreiber, P. Pedrosa, B. Vasconcelos, C. Fonseca, F. Vaz, J. Haueisen, “Modular multipin electrodes for comfortable dry EEG,” *IEEE Engineering in Medicine and Biology Society (EMBC)*, 2016.

[26] P. Pradhapan, R. Griffioen, M. Clerx, V. Mihajlović, “Personalized Characterization of Sustained Attention/Vigilance in Healthy Children,” *eHealth 360°*, pp. 271-281, 2017.

[27] W. Besio, K. Koka, R. Aakula, “Tri-polar concentric ring electrode development for Laplacian electroencephalography,” *IEEE Transactions on Biomedical Engineering*, vol. 53, pp. 926-933, 2006.

[28] K. Koka and W. Besio, “Improvement of spatial selectivity and decrease of mutual information of tri polar concentric ring electrodes,” *Journal of Neuroscience Methods*, vol. 165, no. 2, pp. 216-222, 2007.

[29] S. Patki, B. Grundlehner, A. Verwegen, S. Mitra, J. Xu, A. Matsumoto, J. Penders, and R. F. Yazicioglu, “Wireless EEG system with real time impedance monitoring and active electrodes,” BioCAS, pp. 108–111, 2012.

[30] J. G. Webster, Ed., *Medical Instrumentation: Application and Design*. 4th ed., New York, NY, USA: Wiley, 2009.

[31] C.J. Harland, T.D. Clark, and R.J. Prance. Electric potential probes - new directions in the remote sensing of the human body. *Measurement Science and Technology*, 2:163–169, 2002.

[32] V. Mihajlović, S. Patki, B. Grundlehner, “The impact of head movements on EEG and contact impedance: An adaptive filtering solution for motion artifact reduction*,” IEEE Engineering in Medicine and Biology Society (EMBC)*, 2014.

[33] V. Mihajlović, B. Grundlehner, R. Vullers, “Wearable, wireless EEG solutions in daily life applications: what are we missing?” *IEEE Journal of Biomedical and Health Informatics*, vol. 19, no. 1, pp 6-21, 2015.

[34] J. Xu, B. Busze, C. Van Hoof, K. Makinwa, R. F. Yazicioglu, “A 15-channel digital active electrode system for multi-parameter biopotential measurement” *IEEE Journal of Solid-State Circuits*, vol. 50, pp. 2090-2100, 2015.

[35] A. Palumbo, F. Amato, B. Calabrese, M. Cannataro, G. Cocorullo, A. Gambardella, P. H. Guzzi, M. Lanuzza, M. Sturniolo, P. Veltri, P. Vizza, “An Embedded System for EEG Acquisition and Processing for Brain Computer Interface Applications,” *Wearable and Autonomous Biomedical Devices and Systems for Smart Environment*, pp. 137-154, 2010.

[36] Dreem, https://dreem.com, accessed on March 25th, 2018.

[37] Hitachi WOT-100, http://origin.hitachi-hightech.com/products/images/11484/WOT-100_EN.pdf

[38] Artinis Brite23, https://static1.squarespace.com/static/54ddd9b0e4b0e11f3685f546/t/59772e84db29d6ba1e609051/1500982919314/Brite+Leaflet.pdf

[39] Integrated NIRScap, http://nirx.net/nirscaps/

[40] U. Ha et al., "27.2 A 25.2mW EEG-NIRS multimodal SoC for accurate anesthesia depth monitoring," *Digest of ISSCC*, pp. 450-451, 2017.

[41] R. Pagano, et al., “Improvement of sensitivity in continuous wave near infra-red spectroscopy systems by using silicon photomultipliers.” *Biomedical Optics Express* vol. 7, no. 4, pp.1183–1192, 2016.

[42] P. Schönle, S. Fateh, T. Burger and Q. Huang, "A power-efficient multi-channel PPG ASIC with 112dB receiver DR for pulse oximetry and NIRS," *IEEE Custom Integrated Circuits Conference (CICC*), pp. 1-4. 2017.

[43] Holder, D.S. “Electrical impedance tomography (EIT) of brain function,” *Brain Topography*, vol. 5, no 2, pp. 87–93. 1992.