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THE CHALLENGING SCALP–ELECTRODE INTERFACE AND THE EVOLUTION OF MATERIALS AND ELECTRODE INTEGRATED ICTs FOR ELECTROENCEPHALOGRAPHY

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Electroencephalography (EEG) is a non-invasive technique used to measure the electrical activity of the brain. The use of EEG is very important for the diagnosis of traumatic brain events and mental states such as injury, stroke, depression and many others including the COVID-19 brain fog syndrome. The quality of EEG signals largely depends on the nature of the interface between the surface of the electrode material and the surface of the scalp from where the electrical brain signals are acquired. The scalp surface is composed of an epidermic substrate with hair, grease, dirt, dandruff, skin peels and eventually many different hair products. The electrodes must combine several properties including electrical conductivity, mechanical strength, biocompatibility and corrosion resistance. They also must be manufactured with shapes designed to overcome the inherently complex nature of the scalp–electrode interface. This review reports the latest advances in the design of materials, surface coatings, conductive gels and information and communication technologies being developed to increase the quality of measurement of brain electrical signals in EEG protocols.

Keywords: EEG; dry electrode; substrate; conductive material; gels and creams; tattoo sensor; internet of things.

1. Relevance of Electroencephalography

Electroencephalography (EEG) is a non-invasive technique used to study the electrical brain activity. Electrical brain signals are the consequence of the synaptic activity between neurons.[1] This physiological activity generates different types of electrical signals in the scalp. The EEG technique consists of the detection, recording and interpretation of such signals.[2–5]

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EEG can be used to diagnose brain disorders such as epilepsy, brain tumors, brain damage from head injury, brain dysfunction, encephalitis, stroke, sleep disorders and the identification of mental states such as depression and exhaustion, among others.[6–12] It can also be used to find the right dose of anesthesia to be administrated in patients under medically induced coma and to confirm brain death in patients under deep coma.[10] Recently, during the pandemic caused by the SARS-CoV-2 virus (COVID-19), the EEG has been used to regulate sedation levels in patients in intensive care units and to diagnose and study the so-called "COVID brain fog syndrome".[12]

2. Clinical Limitations of EEG

EEG has several limitations related to the discomfort of the patient during the measurements of the brain signals and to the difficulties in obtaining brain signals with low noise level.[13–15] The material and shape design of the electrodes, the condition of the scalp and the contact efficiency between the electrode and the scalp are the main variables that determine both, the level of comfort and the signal quality of the recordings.[14] Abundant and greasy hair and the presence of dandruff, peels, dirt, conditioners, sprays, styling gels and other hair products may hinder good electrical contact of the electrodes with the scalp.[16]

Electrodes must, somehow, be in excellent electrical contact with the scalp as a primary condition to obtain brain signals of good quality. The electrodes are typically connected with wires to an amplifier and an analog-to-digital converter device. They allow the brain electrical signals to be processed and visualized in a computer.

EEG tests typically take from 20 min to 30 min.[17,18] Testing may require sleeping, opening and closing the eyes, reading a paragraph, looking pictures or flashing lights, breathing deeply, etc.[19–25] Video of body motions is recorded to complement the study during EEG tests in patients with seizures, dementia, epilepsy or Alzheimer's disease.[26,27] Ambulatory EEG tests facilitate longer monitoring outside hospitals and medical facilities. These tests can record brain activity over several days and increase the chances of detecting abnormal electrical brain activities.[28] In all cases, both the comfort of the patient and appropriate electrical contact between the scalp and the electrodes must be ensured.

The EEG signals are classified in five frequency bands according to the EEG Signal Band Division.[19] Each band can be associated with a defined mental state. Delta (0–4 Hz) for sleepiness and dreaming, Theta (4–8 Hz) for drowsiness and meditation, Alpha (8–14 Hz) for relaxed and reflective states, Beta (14–30 Hz) for alert and working states and Gamma (30+ Hz) for active thought states.[19]

3. The Scalp and the Challenging Scalp–Electrode Interface

The scalp is composed of five layers. The outer layer is the skin from which head hair grows. It contains a number of sebaceous glands, hair follicles and is receptive to dirt, sweat, dryness, dandruff and hair products. The second layer is the connective tissue responsible for mechanical properties.[29] It is a dense subcutaneous layer of fat and fibrous tissues containing the nerves and vessels of the scalp. The third layer is the epicranial aponeurosis. It is a fine yet tough layer that connects the frontalis and occipitalis muscles. The fourth layer is the loose areolar connective tissue. This layer provides an easy plane of separation between the upper three layers and the pericranium. The loose areolar tissue is made up of random bundles of collagen I and collagen III. Finally, the pericranium is the periosteum of the skull bones and provides nutrition to the bone and repair capacity.[30,31] Scalp thickness and electrical conductivity may vary from 3 mm to 8 mm and from 0.3 S/m to 0.55 S/m, respectively, depending on factors such as anatomical location, age and sex.[32]

Figure 1 shows schematically the complexity of the scalp–electrode interface. Clearly, the gap between the surface of the electrode and the surface of the outer skin layer is very complex and constitutes a real challenge for ensuring good electrical contact during the EEG testing. The scalp–electrode interface can be understood as an insulating material made of air and hair. Hair is an electrical insulating material and may occur in different types and densities. Additionally, grease, dirt, dandruff, skin peels and different hair products also contribute to the insulating capacity of the scalp–electrode interface. Moreover, there is always a relative movement between the electrodes and the scalp.

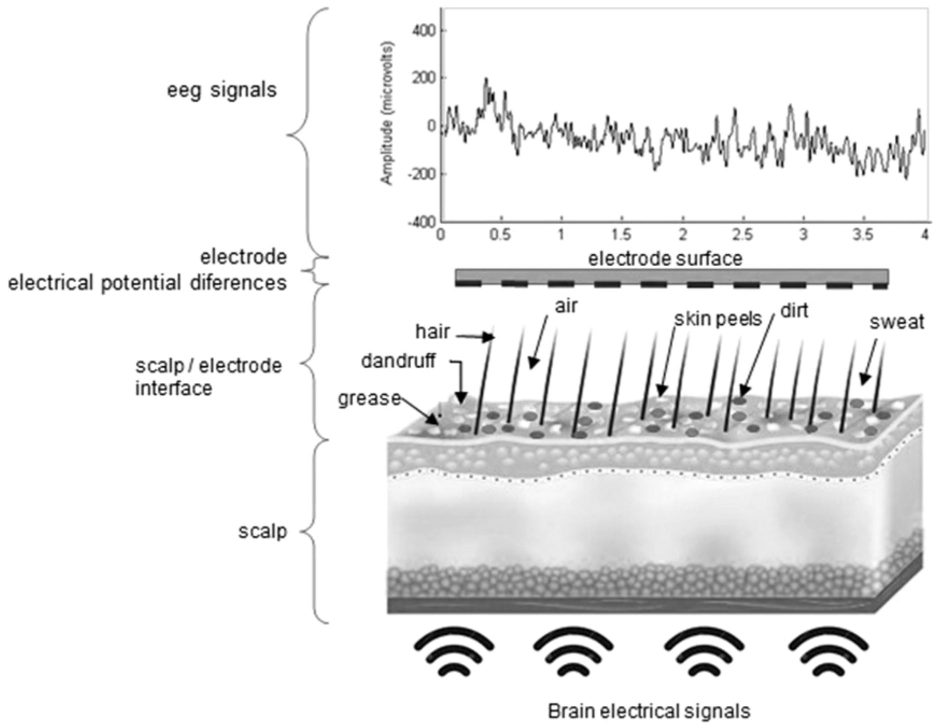

Fig. 1. Cross-section of the scalp and underlying skull at the vertex.

Several types of electrodes have been designed to overcome the complex nature of the scalp–electrode interface. Electrodes can be classified according to the strategy they use to ensure electrical conductivity at the scalp–electrode interface.

4. Wet Electrodes

The most commonly used wet electrodes are constituted by an Ag/AgCl electrode piece supported in a snap connector made of rigid metal alloys or semi-rigid conductive plastics in the form of discs or cups. Common metal snap connectors are made of titanium (Ti), gold–palladium–rhodium (AuPdRh), stainless steel (SS316) and platinum–rhodium (PtRh)-based alloys.[33] Common plastic snap connectors are made of polyacetylene, polyphenylene vinylene, polypyrrole, polythiophene and polyaniline.[34]

Wet electrodes employ conductive gels or creams to fill up the gap occurring between the electrode and the skin.[35] These ionic conductive media improve the signal registration by ensuring good electrical conductivity between the surface of the skin and the electrodes.[36] They lead to obtain impedance values between 1 and 10 kΩ.[11] This range allows the detection and recording of electrical brain signals with high signal-to-noise ratio and enables accuracy in further signal processing and studying.[11,33,36] Table 1 shows the composition and electrical conductivity of several commercial gels and creams used in EEG studies.[37]

Another advantage of wet electrodes is their high level of comfort during the EEG measurement. Wet electrodes are completely painless even during long periods of time.[16] However, the process of removing the conductive gels or creams from the scalp after each EEG acquisition session can be a tedious and unpleasant process. Conductive gels and creams tend to stick to the hair and run off over the skin. In most of the cases, absorbent towels, polar solvents, such as water and ethanol, and a final hair wash are required to remove the gels and creams from the scalp.

5. Dry Electrodes Based on Rigid Multi-Pins Arrays

In recent years, researchers have shown much interest in dry electrodes based on multi-pin arrays. Dry electrodes do not require the use of conductive media nor any skin preparation. They are made of

Table 1. Composition and conductivity of commercial gels and creams for EEG studies.

Name	Composition	Conductivity (S/m)
Cogel by Comedical	Water, propylene glycol, sodium carbomer, lithium chloride, disodium ethylenediaminetetraacetic acid, methyl paraben	2.02
Neurgel by Spes Medica	Water, potassium chloride, hydroxyethylcellulose, propylene glycol, methylchloroisothiazolinone, methylsothiazolinone, bensyl alcohol	4.18
Onegel by Ates Medica	Not declared	0.37
Zerogel by Eurocamina	Not declared	0.30
Accream by Spes Medica	Water, polyethylene glycol (25), cetyl/stearyl ether, glycerin, hydrated silica, calcium carbonate, potassium chloride, hydroxyethylcellulose, sodium chloride, phenoxyethanol, ethyl methyl propyl butyl para-hydroxybenzoate (paraben), propylene glycol	1.41
EC2 by Grass AstroMed	Potassium chloride, sodium chloride and the rest chemical composition is not declared	2.35
Ten 20 by Weaver and Co	Water, polyoxyethylene 20 cetyl ether, glycerin, calcium carbonate, propylene glycol, potassium chloride, gelwhite, sodium chloride, polysorbit 20 sorbitol, methyl propyl paraben	0.54
Bionen by Ates Medica	Not declared	0.063
Onecream by Ates Medica	Not declared	0.088

conductive materials and thus they are directly applied to the scalp.[38,39] These electrodes can be used by non-expert technicians and facilitates their use outside medical centers.[40,41] The multi-pin arrays are designed to overcome the limitations for efficient electrical contact due to the presence of hair, dirt, sweat, dryness, dandruff and hair products in the gap between the surface of the electrodes and the scalp. The pins are directly applied to the scalp by exerting some pressure on them. The number of pins, the way they are supported, the level of exerted pressure, the conductivity of the material and the conditions of their surfaces, determine the quality of the electrical contact between the electrode and the scalp. The effect of the applied force on the performance of dry multi-pin electrodes is a matter of study.[42] In general, the higher the applied force, the lower the impedance of the electrode–scalp interface, the better the quality of the acquired signal and the higher the discomfort of the patient. The discomfort of the patient find a limit when became painful.[42]

Rigid multi-pin electrodes are very complex in shape and their manufacture generally requires the use of thermoplastic materials. Such materials yield multi-pin arrays by the injection molding technique. There are several strategies to produce rigid multi-pin electrodes by injection molding.[4,15,43] Injection molding of conductive polymers such as polyacetylene, polyphenylene vinylene, polypyrrole, polythiophene and polyaniline. Injection molding of non-conductive polymers, such as polyethylene and polystyrene, blended with conducting powder materials such as silver or copper. Injection molding of non-conductive polymers and subsequent coating with conducting paints can be mentioned among the most common approaches to produce rigid multi-pin dry electrodes. The improvement of the electrical properties of powder materials is also a matter of research.[44] The resulting body of the multi-pin array must be chemically inert to prevent electrochemical noise and degradation of the brain signal quality especially when the electrode is in contact with sweat.[45]

Dry electrodes usually present higher electrode-scalp impedance than wet electrodes. Dry electrodes impedance ranges between 50 and 256 kΩ.[16,46] *In situ* active amplification of registered brain signals is generally required.

6. Dry Electrodes Based on Spike Arrays

Dry spike electrodes were developed in an attempt to diminish the impedance of the electrode–scalp interface. They are based on the penetration of the epidermis by micro-needles.[47] This type of electrodes has direct contact with the electrically conductive epidermis and converts brain signals into electronic currents in the electrode. In general, spike electrodes are usually made of conducting tips with sharpened endpoints. The rest of the tip is covered with insulating polymers.[48]

Several methods are reported for the fabrication of micro-needles. An extended technique is the so-called replica-molding process. In this technique a negative master mold made of brass is micro-machined by a computer numerical control milling machine according to the desired dimensions. Then, a resin is poured in the mold and cured. The piece is removed from the mold and a silver or gold film followed by a parylene film are deposited by chemical vapor deposition technique.[49]

Another extended method is lithography. Here, a photo-resistant polymer layer is deposited over a glass surface by spin coating technique. This layer is subsequently cured by exposure to UV light in front of a lithography mask aligner. The micro-needle array photomask is then brought into contact with the backside of the glass and the UV exposure is made from the backside on the mask aligner. The final step consists of the deposition of a thin coat of silver by electron-gun evaporator. This fabrication procedure gives the possibility of having the micro-needle arrays formed on flexible substrates.[50]

The thermal drawing method is also of interest. First, a thermoplastic film is fixed to a copper block and heated until its glassy state. Second, another copper block with a pillar array is heated at slightly higher temperature. Third, the copper bock with pillar array is moved towards the glassy-state polymer film until touching it and immediately pulled back at a defined speed. Then necks are formed between the pillars and the polymer film due to the surface tension. Fourth, the polymer film is rapidly cooled by refrigeration of the copper block. The pillar array is moved back continuously and the necks are broken forming a micro-needle tip array.[51]

Finally, micro-needles arrays can also be produced by the formation of extremely thin scaffolds by chemical growing of silicon material followed by platinum or titanium metallization by sputtering technique and final parylene deposition.[52]

In all cases, spike electrodes are medical disposable materials and have an intrinsically higher risk of infection than wet and multi-pin-based electrodes.

7. Dry Active Electrodes

An alternative way of increasing the signal quality of dry electrodes is to incorporate microelectronic devices into them. Microelectronic devices can be designed for a number of functions and permit to expand the limits of collection of EEG data. This type of electrodes is called dry active electrodes. The incorporation of microelectronic devices is possible by the use of electronic miniaturization techniques.[53]

Exemplary, a new dry active electrode is reported to incorporate a bio-potential amplifier and an inertial measurement unit containing an accelerometer, a gyroscope and a magnetometer. Both devices are connected to a microboard capable of processing the analogue signals and to another microboard for controlling the conditioning circuitry, collecting and storing the brain signal data. All systems are powered by a single battery.[54]

Dry active electrodes also admit ergonomic designs. Exemplary, flexible and stretchable dry active electrodes for multi bio-potentials sensing based on electrically conductive composites are reported.[55] This ergonomic dry active electrode consists of a soft substrate, an electrode and simple circuits including a microamplifier and a microcapacitor. The soft substrate is made of molded silicone, the electrode is fabricated by coating the patterned substrate and the circuits are interconnected each other by soldering components using flexible connections.[55] Another aspect of ergonomics is the possibility of wireless connectivity in EEG studies.[56]

Dry active electrodes can also be used to reduce the noise level of the registered brain signals. In this case, the microelectronic device installed in the electrodes can be designed to reduce the noise of transferred signals from the electrode to the acquisition systems and/or reduce the electromagnetic interference coming from the surrounding environment.[57,58]

8. Dry Electrodes Based on Flexible Multi-Pin Arrays

A serious limitation of dry electrodes based on rigid multi-pin and spike arrays is the discomfort caused by the tips of the pins or spikes when pressed on the scalp. When the EEG session takes a long time, the pressure of the pins and spikes on skin may become quite painful.[43]

Dry electrodes based on flexible multi-pin arrays were developed in an attempt to reduce the discomfort of patients during EEG sessions. One of the first dry electrodes with flexible multi-pin arrays were developed using polyurethane as thermoplastic flexible supporting material and titanium and titanium-nitride-based conductor coatings. This study reported the effect of the applied force on the values of the impedance between the electrode and the scalp. The registered impedance was below $200\,\mathrm{k}\Omega$.[41] In a subsequent attempt, a modular multi-pin electrode using a polyurethane-based electrode coated with conductive paint composed of titanium, titanium-nitride, silver chloride and gold powders was built. These electrodes yielded an impedance below $150\,\mathrm{k}\Omega$. Clearly, dry electrodes based on flexible multi-pin arrays could lead similar impedance values than wet electrodes.[41,46,59]

More recently, dry and flexible electrodes based on silicon were investigated. Silicon was blended with conductive copper and iron powders. The resulting impedance values ranged from $4\,\mathrm{k}\Omega$ to $14\,\mathrm{k}\Omega$. These values are acceptable for standard EEG measurements. Finally, they showed stable impedance even during long EEG sessions.[15]

Flexible electrodes using polyimide (PI) as substrate material were also investigated. They offered flexibility and comfort in contact with the skin. The substrate material was coated with a conducting polymeric material based on poly(3,4-ethylenedioxythiophene) (PEDOT) doped with poly(styrene sulfonate) (PSS). PEDOT:PSS/PI electrodes recorded brain activity in a very comfortable and non-invasive manner.[60,61]

The use of polydimethylsiloxane (PDMS) as substrate to build EEG electrodes is also reported. PDMS is a flexible, biocompatible and non-toxic polymer. The electrode was based on slender and soft pins able to get through the hair in order to reach the scalp surface. Sputtering process allowed to deposit a gold layer on the PDMS surface of the electrode. This electrode showed higher impedance than standard wet electrodes without skin preparation.[4,62]

Interestingly, a new concept based on silver-ball (Ag/Ball) is investigated. Ag/Ball of 1.5 mm of diameter was embedded in PDMS and connected via a $50\,\mu$m copper wire to an acquisition signal device. The Ag/Ball-PDMS electrode had an impedance of $10.6 \pm 2.7\,\mathrm{k}\Omega$, which are higher than the impedance of the wet electrode used as reference ($2.8 \pm 0.4\,\mathrm{k}\Omega$). However, the Ag/Ball-PDMS electrode was reported to record high quality EEG signals.[63,64]

9. Adhesive Dry Electrodes

In 2010, the first adhesive sensor for measuring brain alpha waves was reported.[65] This sensor consisted of a non-invasive stretchable electrode array. The electrode was made of several layers of PI and poly-methylmethacrylate (PMMA). This substrate was used to encapsulate chromium, gold and other conductor materials. The adhesive electrode registered impedance values from $15\,\mathrm{k}\Omega$ to $60\,\mathrm{k}\Omega$.[65]

Later, several adhesive flexible electronic sensors based on graphene coated with an ultra-thin layer of transparent PMMA were developed. Graphene has excellent properties such as intrinsic flexibility, reliable electrical performance and high chemical stability. Graphene-based adhesive sensors are characterized by their high electrical and mechanical performances. These adhesive sensors were able to detect alpha rhythm with a peak at 10 Hz and registered signals of similar quality than wet electrodes.[66–69]

The use of graphene became widespread rapidly. Adhesive electrodes have been fabricated by semi-embedding highly graphitized electrospun fiber/monolayer graphene into a soft elastomer. In a demonstrative study, the electrodes were stuck to the forehead of patients to detect brain alpha waves. Reference electrodes were placed near the earlobes. EEG signals were recorded while the subjects rested in relaxed and motionless status with the eyes closed for a period of 30 min. The EEG recordings detected signals around 10 Hz, which are characteristic of alpha waves.[21,25]

The fabrication of adhesive electrodes based on conducting polymer blends was also reported. These polymer blends were composed of PEDOT:PSS. The polymer blends were deposited by spin-coating or inkjet printing, on the top of a commercially available temporary tattoo paper. The tattoo paper was adhered to the skin by means of Van Der Walls forces. The tattoo paper was composed of decal transfer and glue. Decal transfer worked as the substrate of the electrode and the glue sheet provided adherence to skin while preventing short circuits between interconnection lines. It also included a PI film used as supporting layer for the external electrical connection. The validation of this adhesive electrode was carried out by monitoring alpha waves in a patient. Alpha waves were detected around 10 Hz during one minute of relaxation with closed eyes.[26–30]

The use of composites made of carbon nanotubes (CNTs) and PDMS elastomers to construct electrodes for EEG is also attracting the attention of the scientific community. The CNT/PDMS composites could be prepared by sonication of the components dissolved in isopropyl alcohol (IPA) at 40 kHz frequency and 80–100 W of power during 10 min. After complete vaporization of the alcohol the resulting CNT/PDMS composites showed high electrical conductivity with resistances $< 20\,\Omega$/sq, tensile strength of $\sim$3.65 MPa, flexibility of more than 90°, elasticity of more than 45% yield strain and excellent sensitivity and stability properties. Alpha-rhythm waves in the band of 8–12 Hz were registered for CNT/PDMS sensors stuck on the right forehead of a subject in a relaxed state.[66,70] The use of CNTs combined with conductive and magnetic nanoparticles is also been explored in an attempt to enhance the sensitivity of sensors and electrodes.[71]

Adhesive electrodes can adhere to hairy scalps for long periods. Thus, they are recommended for long EEG studies. However, they present some serious limitations related to discomfort, scalp irritation, allergic skin reactions, impedance variation due to long time of recording and serious difficulties to detach the electrode from hairy scalps.[72]

Figure 2 shows schematically, the conceptual differences between wet, rigid multi-pin, flexible multi-pin and adhesive EEG electrodes.[73] Table 2 shows a comprehensive list of polymers and conductive particle sets reported as materials intended to be used in the fabrication of dry electrodes based on multi-pin, spike and adhesive sensors.[43,74]

10. ICTs in the EEG Electrode Development

Information and communication technologies (ICTs) is an extensive term for information technology that emphasizes the role of unified communications. The integration of telecommunications, computers, software, hardware, data storage and audiovisual systems allow users to access, store, transmit and manipulate information very fast and easy. ICTs become an important part of the EEG electrode designs and permit to overcome many limitations imposed by the nature of the materials they are made of. In particular, ICTs intervene with associated software and hardware for brain signal amplification, signal noise reduction and connectivity.

The EEG amplifier is the part of the data acquisition system responsible for accommodating, amplifying and converting the analog brain signals from the sensing electrodes into digital signals that can be processed by a computer. Most of the advances in EEG amplifiers are related to improvements in the sampling rate, bandwidth definition and resolution.[83] The sampling rate has been increased considerably during last years. The sampling rate describes the number of times that the analog brain signal is measured per unit of time to be converted into a digital signal. The higher the sample rate, the better the quality of the resulting digital signal and the higher the amount of data to be acquired, processed, transmitted and stored. The other aspect of permanent improvement is the bandwidth. The bandwidth is the effective frequency band that the EEG system can measure. The bandwidth is defined by the quality of frequency filters. Note that these filters are designed to attenuate the frequencies below and

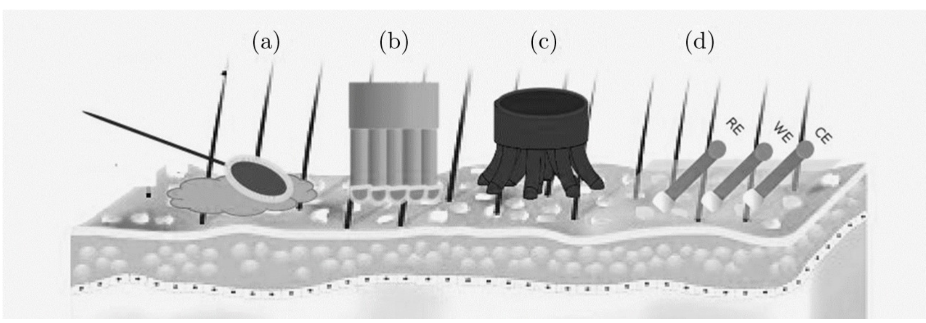

Fig. 2. Different types of electrodes for EEG tests: (a) wet electrode; (b) dry electrodes based on rigid multi-pin arrays; (c) dry electrodes based on flexible multi-pin arrays and (d) adhesive dry electrodes.

Table 2. Substrate and conductive materials for fabrication of dry electrodes.

Polymer	Conductive material	Design	Impedance	Measurement conditions	Key features of electrode	Ref.
PU	TiN	Multi-pin array. 24 conic pins. Pin: $h = 6$ mm; $\Phi = 1$ mm.	65–118 kΩ	No skin preparation. Compared against electro-gel.	PU provides contact reliability and patient comfort. TiN is biocompatible and electrochemically stable in contact with human sweat.	59
PDMS Parylene C	Au	Multi-pin array. 13 and 21 pins. Pin: $h = 5$ mm, $\Phi = 1$ mm.	< 20 kΩ	Skin was cleaned with alcohol pad. Compared against conventional wet electrode.	PDMS provides flexibility and chemical and thermal stability, Parylene C film provides binding force between PDMS and Au film.	4
PB	Ag	Multi-pin array. Pins: 10 mm long.	80 kΩ	No skin preparation. Compared against commercial gel.	Flexible conductive bristles based on Ag particles instead of pins to avoid the force applied to reduce impedance and to obtain a high signal quality.	75
Resin	Conductive paste	Multi-pin array. Pin: $\Phi = 1.25$ mm, $h = 11$ mm.	< 1 kΩ	No skin preparation. Compared against wet and dry commercial electrodes.	Low cost of fabrication using a 3D printer.	76
	Au/Be/Cu	Multi-pin array. Probe head with spheroid radius of 1.3 mm.	4–14 kΩ	No skin preparation in the dry electrode site. Skin preparation in the wet electrode site.	The probe has $\Phi < 0.5$ mm to avoid pain when contact with the scalp.	15
PU	Ag/AgCl	Multi-pin array. Pin: $\Phi = 1$ mm, $h = 6$ mm.	< 150–164 kΩ	Skin was cleaned with ethanol and a soft cloth. Compared against commercial Ag/AgCl gel.	PU is flexible, moldable and cost-efficient for production.	77
TPU	Ag/AgCl	Multi-pin array. Pin: $\Phi = 2$ mm, $h = 6$ mm.	38.6 kΩ	No skin preparation. Compared against commercial Ag/AgCl gel.	TPU provides flexibility and makes it comfortable to wear.	78

Table 2. (*Continued*)

Polymer	Conductive material	Design	Impedance	Measurement conditions	Key features of electrode	Ref.
RS	Cu–PI–Cu	Contact pad. $4\,\text{mm} \times 4\,\text{mm}$ rectangle shape. $h = 4\,\text{mm}$.	$< 10\,\text{k}\Omega$	No skin preparation. Compared against commercial Ag/AgCl wet.	RS and Cu–PI–Cu sheets provide comfort and good contact with the skin.	79
PMMA	Ag	Multi-pin array. Pin: $\Phi = 1\,\text{mm}$, $h = 2\,\text{mm}$.	$50\,\text{k}\Omega/\text{cm}^2$	No skin preparation. Compared against commercial Ag/AgCl gel.	Low skin-contact impedance and susceptibility to acquire high quality biopotentials.	80
FPCB PDM	Ag–AgCl	Multileg Leg length $= 7.0\,\text{mm}$ Leg length $= 3.5\,\text{mm}$	$< 150\,\text{k}\Omega$	No skin preparation. Compared against wet electrode.	Flexible and deformable with low skin-low-contact-impedance with the scalp	81
PVA tattoo paper	PEDOT PSS	Tattoo sensor. $\Phi = 12\,\text{mm}$	$16.7\,\text{M}\Omega/\text{cm}^2$	No skin preparation. Compared against Ag/AgCl electrode.	Tattoo sensors adhere to skin generating imperceptible contact between the scalp and the electronics.	82
PMMA-PI tattoo paper	Graphene	Tattoo sensor based on filamentary serpentine shapes.	$< 20\,\text{k}\Omega$	No skin preparation. Compared against commercial Ag/AgCl gel.	Graphene provides a good adhesion and imperceptible contact to the skin.	67

Notes: PU: Polyurethane; PDMS: Polydimethylsiloxane; PB: Polymer Brush; TPU: Thermoplastic polyurethanes; RS: Rubber silicon; PMMA: poly(methyl methacrylate); PVA: Polyvinyl alcohol; PI: Polyimide; TiN: Titanium Nitride; Au: Gold; Ag: Silver; B: Beryllium; Cu: Copper; Cl: Chloride; PEDOT: poly(3,4-ethylenedioxythiophene); PSS: poly(styrene sulfonate).

above the bandwidth. The better the filters, the lower the noise in the resulting signal. The resolution of the amplifier is the number of bits that the analog-to-digital converter of the amplifier uses to code each analog voltage value into a digital voltage value. The higher the number of bits the higher the resolution of the amplifier.[84]

EEG recordings are often contaminated by different kinds of noise. Noise is anything the electrodes detect without the intention or the purpose to be detected. There are three different sources of noise in EEG data. Physiological noise includes cardiac signals and artifacts caused by muscle contraction, signals caused by eyeball movements and irrelevant underlying brain activity not pertaining to the experiment. Environmental noise is anything that uses electricity and emits an electromagnetic field that is detected by the electrodes. Some examples of such external noise sources are the alternating current power lines, room lighting and electronic equipment in the vicinity of the sensors. Finally, motion artifacts are the noises caused by the movement of any physical part of the measurement setup. These movements can happen at the intersection of electrodes or when moving electrode cables.

All these annoying signals limit severely brain recording utility and, hence, have to be removed. The cleaner the data, the more representative and accurate the analysis. Current strategies to increase the signal-to-noise ratio are based on the increase of the sample size and averaging and cleaning the raw data.

The strategy for increase of the sample size and averaging is supported under the hypothesis that random noise is averaged out in bigger sample sizes with ultimate signal prevailing.[83] However, for long recording sessions or when participants are difficult to recruit, data cleaning may be a better option.

Data cleaning can be carried out before, during and after EEG recording. There are several simple and effective measures taken before EEG recording that certainly will reduce the noise in the acquired raw data. Perform the experiments in electromagnetically isolated environments, remove or replace any electronics that use alternating current with equipment using direct current, ensure the participants are in a comfortable resting position to reduce motion artifact noise, eliminate physiologically induced noise by removing tasks that require verbal responses or large movements and shorten recording sessions can be exemplarily mentioned. Noise occurring during EEG recording is mainly caused by the movement of electrodes and cables. Minimization of moving parts includes the use of minimal cable lengths. The less the cable length, the better. Finally there are many mathematical tools to increase the signal-to-noise ratio after EEG recording. These post-processing techniques can be classified into two groups: those designed to identify and separate the signal from the noise with the purpose to further elimination of noise and those designed to eliminate the noise by auto-rejection.

Artifact Subspace Reconstruction is an online, component-based method to effectively remove transient or large-amplitude artifacts. The technique is capable of running in real time and uses statistical anomaly detection to separate artifacts from EEG signals in multichannel data sets. It assumes that non-brain signals introduce a large amount of variance to the data set and can be detected via statistics.[85,86]

Independent Component Analysis is a technique that effectively decomposes the multichannel EEG data into multiple independent components belonging to either artifacts or neural sources. It is built on the observation that artifact and neural signals possess distinguishable spatio-temporal patterns.[87,88]

Canonical Correlations Analysis uses autocorrelations within a given time series to characterize the signal component of an unclean data set. Autocorrelation is the correlation between a signal and a lagged copy of itself over successive time intervals.[89,90]

Multiple Sparse Priors uses Bayesian statistics and has outperformed conventional methods in some studies.[91]

Auto-rejection approaches require working with high density EEG systems. These are designed to automatically detect and reject "bad" channels in real time and to avoid incomplete data arrays by the interpolation of "bad" channels using signal sets of "good" neighboring channels.

Sensor Noise Suppression method assumes that true signals are picked up by more than one sensor. As such, each channel is projected onto the subspace spanned by its neighbors and replaced by its projection. The technique removes the noise by seeing what bits of its signal are unique to a given channel and what bits show up in other channels. Again, the idea here is that brain signals will project onto multiple sensors, while noise is uncorrelated across sensors.[92]

Fully Automated Statistical Thresholding for EEG artifact Rejection uses a combination of five different statistical criteria to identify "bad" sensors. These criteria are variance, correlation, the Hurst exponent, kurtosis and line noise.[93]

Random Sample Consensus uses statistical models to data that contain outliers. Here random subset of sensors (25% of the sensors, called "inliers") is sampled. The data in all sensors are then interpolated from these inlier sensors. This is repeated multiple times resulting in sets of data series for each sensor with reduced noise.[94]

Cross-validation on Signal Data dynamically adapts peak-to-peak thresholds based on the characteristics of the data to detect and reject bad channels. They replace bad data using interpolation from neighboring channels where possible.[95]

Deep Learning uses existing data sets to train an Artificial Intelligence algorithm to automatically detect and reject artifacts from a previously trained recording. Ideally, this is done in real time.[96,97]

Finally wireless connectivity empowered with Internet of Things (IoT) including real-time cloud calculations permits to design wearable multimodal monitoring EEG systems. These systems are able to perform simultaneous monitoring of multichannel EEG, regional cerebral oxygen saturation, body surface temperature, electrocardiogram, photoplethysmography and bioimpedance. The IoT platform and wireless devices enable the collected scalp signals available for remote diagnosis.[98]

11. Future Outlooks

Skin wearable electronics has experienced a skyrocketed scientific interest during the last 5 years.

Technologies tend to prioritize the comfort of the patient. In this aspect, dry and flexible electrodes have a significant advantage over wet, rigid and spike electrodes. Clearly, dry flexible multi-pin electrodes are particularly relevant to EEG. One of the main challenges in the area is the integration of two types of materials, namely: the flexible and shape customizable substrate material and the conductive material capable of transducing electrical signals. Another challenge is the development of an efficient conformation shaping process. The analysis of the scientific literature shows a clear technology trend. Integration by injection molding of elastomers, such as silicon, with microparticulate conducting materials, such as silver and graphene, seems to be the most promising approach.

The need to increase the quality of the acquired brain signal is the driving force for the following three complementary technological areas: the decrease of the impedance of the electrode–scalp interface through better materials and shape designs, the increase of sampling rate, bandwidth definition and resolution through better microelectronics and data cleaning through better post-processing techniques based on mathematical tools.

12. Conclusion

This review summarized the materials used in the construction of surface EEG electrodes. Electrodes were grouped into five types: wet, dry based on rigid multi-pin arrays, dry based on flexible multi-pin arrays, dry based on spike-arrays and adhesive electrodes. The materials employed as substrates consist of metals and polymers such as PDMS, PMM, tattoo paper and PU, among others. These polymers are characterized by its biocompatibility and a Young Modulus similar to human skin. Graphene, Au and Ag nanoparticles, CNT and PEDOT are employed as conductive materials.

Complementary advances in microelectronics and data cleaning were also presented. The current main challenges of the technique are the actual instrumentation of the use of these types of devices for daily monitoring, the increase of the signal-to-noise ratio and the maintenance of the electrode stability during its life time.

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Conflict of Interest Statement

No conflict of interest exists. All authors have nothing to disclose.

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