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“wearable EEG system for Auditory Attention Recording”

Deliverable 1.1: “Specification of the functional electrodes”

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Releases history

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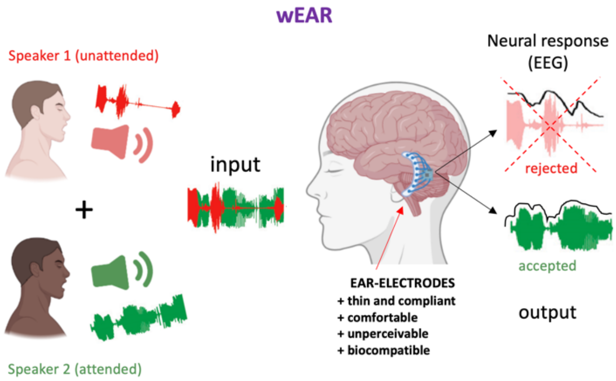
Version 1	20/02/2023		Report on technical specifications for the realization of the wEAR prototype functional electrodes

ABSTRACT

The wEAR prototype, a wearable EEG system for Auditory Attention Recording, is an innovative lab-based hearable (i.e. wearable on-ears) technology that can estimate and decode the wearer's auditory attention, by processing her/his electroencephalography (EEG) signals. In its final shape, it comprises ultra-thin, conductive polymer based electrodes capable to couple with human ears reliably, comfortably, robustly and with aesthetic acceptability for the user, avoiding signal artefacts induced by head movements. wEAR will take advantage of novel conductive polymers dry electrodes, with improved sensitivity, capable of recording for longer periods of time without the need conductive gels. The functional electrodes will be connected with a self-contained module (based on standard electronic components – amplifier, filters, battery etc.) for the EEG data processing – pre-processing (artifacts/noise removal), features extraction and classification (machine learning) – in order decode which source the listener wants to attend to.

With wEAR a new perspective can be given to hearable EEG devices, making them comfortable enough so to be properly embedded in the user's daily life. Furthermore, it will provide a better support to the functionality and rehabilitation challenges that arise for hearing impairment subjects.



Simplified illustration of the wEAR prototype

The main objective of the project is, therefore, to demonstrate a new multi-functional technology that will be, in the future, potentially interfaceable with commercially available hearing aid devices as a complementary sound source localization tool.

This deliverable (D1.1) will review the specifications (electrical and mechanical) of the electrodes that can be relevant for the scope of the project.

D1.1 comprises:

- an overview of the Brain Mapping techniques, with a particular focus on the electroencephalography (EEG);
- a description of the typical electrodes employed for EEG recording, highlighting the pros and cons of each approach;
- then there will be the core of the D1.1, i.e. the description of the specification of EEG electrodes and their quantification;
- finally, the wEAR prototype will be presented as well as the wEAR project workflow.

CONTENTS

1.1 OVERVIEW OF THE EEG AND THE OTHER BRAIN MAPPING TECHNIQUES

The *electroencephalography* (EEG) is a well-known and recognized method for the recording of brain electrical fields, discovered by the German psychiatrist Hans Berger, in 1929. Electroencephalography, together with *Functional Magnetic Resonance Imaging* (fMRI), *Functional Near-Infrared Spectroscopy* (fNIRS) and *Magnetoencephalography* (MEG) are the most important techniques for understanding the information on brain structure and neural activity. They are widely employed for applications spanning from clinical, such as diagnosis and management of patients with epilepsy, to neuroscience. On one hand, fMRI and fNIRS are neuroimaging systems based on the measure of the changes in blood flow – which is correlated with the electrical neurons activity due to neurovascular coupling – to determine brain functions and neural activity. On the other hand, MEG and EEG can map directly the brain activity through (i) the magnetic fields generated from electrical currents occurring in the brain, as for the MEG, or (ii) the electrical potentials caused by the neurons electrical action potential (AP), the so-called neural spike, as for the EEG.

The strength of fMRI and MEG is the highest temporal resolution (for the former) and spatial resolution (for the latter). Despite this enormous advantage, these systems are employed exclusively in clinics and hospitals. Indeed, besides the costs that typically exceed thousands of euros, the size of the equipment does not allow the portability and their use for everyday life. The development of a wearable brain activity monitoring system has been therefore a long chased milestone, since portable tools would enable the daily monitoring of brain neural activity, which is fundamental for the identification of the very moment in which the illness state begins (such as epilepsy, for instance). Portable biomedical devices can really improve the quality of patients life with a reduction of health care expenses.

In this appealing scenario, EEG and fNIRS offer the opportunity to be developed into portable or wearable devices. With respect to fNIRS, EEG is now gaining more attention on the development of wearable neural activities recording systems, due to an higher temporal resolution (to a millisecond range, slightly lower than that achievable with MEG), higher versatility and cost-efficiency.

The electroencephalogram can monitor human's brain activity through electrodes located on the head of the user. These electrodes should be highly sensitive to the APs coming from a large population of neurons that carries physiological information. Naturally, neural signals are attenuated due to the conductive properties of the tissue layers encountered from the source (brain) to the recording system. Indeed, it is well-known that *skin*, *skull*, *dura mater*, *cerebrospinal fluid* (CSF), and pia mater, play a key role in the quality of the measurable EEG signal. For instance, CSF has a not negligible effect, called *spatial blurring*, of the EEG signals, which can be reduced by changing the head position or movement. On the other hand, the recording of brain signals from electrodes placed on the surface of the skin relies critically on the electrode-skin interface. Indeed, the skin is constituted of different layers – the *subcutaneous tissue* (which connects the dermis with the other underlying tissue), the *dermis* and the *epidermis* (or *stratum corneum*) – and, as will be reported in the next Sections, the electrode-stratum corneum interface have to be engineered and optimized in order to provide good and constant contact, thus reducing as much as possible the contact impedance and, consequently, the signal attenuation/loss. To this end, it is of paramount importance the materials and type of electrodes selection.

1.1.1 CATEGORIES OF EEG ELECTRODES

Wet electrodes

The working principle of wet electrodes is based on a conductive gel/paste that is placed between the electrode and the skin. The role of the gel is fundamental for the minimization of the contact impedance at the *electrode-tissue interface* (ETI), since it act as a bridge for the ionic current that flows from the skin, and the electronic current that flows to the electrode. Typically, wet electrodes are based on (i) non-polarizable silver–silver chloride (Ag/AgCl) metal surrounded by a hydrogel that contains chloride, or on (ii) others polarizable metals (especially gold – the so-called *gold cup*). For these systems, the electrode–tissue interface can be accurately modelled and schematized as a series of two RC circuits, two resistors, and two potential sources, as shown in Fig. 1 (a). The RC circuits grasp the complex resistive and capacitive phenomena at the electrode–electrolyte ($R_{ee}//C_{ee}$) and at the skin–electrolyte ($R_{se}//C_{se}$) interface. R_g and R_t correspond to the resistance of the electrolyte and the skin, respectively. The potential source E_{hc} and E_{se} corresponds to the potential difference between electrode–electrolyte and skin–electrolyte, respectively. Wet electrodes are widely recognized as the "gold standard" for the recording of biopotentials, and in particular EEG, for both clinical and research applications. The use of the gel increases adhesion of the electrode to the skin, thus guaranteeing a stable contact at least until the conductive properties of the interface are stable (i.e. before that the electrolyte starts to dry out). Indeed, when the conductive gel/paste is drying, it can significantly affect and reduce the signal quality, thus limiting the electrode useful lifetime to, at most, a few hours. Moreover, the use of EEG recording systems based on wet electrodes requires expert supervision: the setup process takes time and effort since is fundamental (i) a proper preparation the skin (which should be clean) before the application of the conductive gel/paste, (ii) the correct positioning the electrodes (or the cap) and (iii) to check that the EEG signal quality level is acceptable and meaningful. Another important drawback of wet electrodes is related to the skin irritation caused by the gels and pastes that contact the skin for a prolonged time. For these reasons, EEG recording systems based on wet electrodes are therefore inconvenient for daily-life applications, and it is now well accepted that a gel-free system is preferable for long-term (i.e. > few hours), out-of-lab, on skin recording.

Dry electrodes

Dry electrodes do not need the use of any conductive gel/paste, thus making the transduction from the ionic current at the biotic side (tissue) to the electronic current at the abiotic side (electrodes) much more complicated. They exploit unique shapes and materials to improve current transduction: typically, dry electrodes are based on gold-coated small pins in order to reduce different impedance sources (such as that caused by the hair) and improve the electrode conductivity. In some conditions, pins-electrodes performance can be comparable to that of Ag/AgCl wet electrodes. From the electronic point of view, the absence of conductive gel/paste at the biotic-abiotic interface, results into an obvious increase of impedance and higher capacitive components. The electrical modelling of the dry electrodes ETI must take into account of the above mentioned behaviours. The capacitive and resistance components related to electrolyte (C_{ee}, C_{se}, R_g) should be removed from the model, and substituted by using a (i) non-physical component, the constant phase element CPE_{se}, which describes the *electrical double layer* (EDL), and a (ii) *Warburg element* (Z_w) that grasps the ion diffusion at the biotic-abiotic contact interface. The model of the dry electrodes ETI is reported in Fig. 1 (b).

As already mentioned, the use of a conductive gel-free systems are preferable for long-term on skin recording. However, there are still severe limitations in terms of performance that have largely precluded their wide diffusion for every-day life uses. Indeed, since EEG signals are typically small – in the range from few to tens of µV due to their propagation through the head volume – these systems are considered to be too noise prone to be used in daily life applications, especially because of the weak reliability of electrode-skin connections. Therefore, a proper engineering of the ETI is the first step to face out. Then, depending on the metal employed for dry electrodes, it has been reported that they are susceptible to (i) half-cell potentials created by sweat that can skew signals, and/or to (ii) the so-called *movement artefacts*, thus leading to severe issues especially for daily, out-of-lab applications. The employment of metal electrodes can leads to a high power dissipation, signal degradation and they required high stabilization time (i.e. the time that is needed before producing a stable output). Furthermore, forces applied to put in contact the electrodes and the skin (such as headset) are typically not reproducible, thus producing too high variability, especially in terms of frequency response that can be unstable over time. Last but not least, from the structural point of view, the mechanical mismatch between the the metal electrodes (typically bulky and brittle) and the skin (typically soft) could leads to micro-fractures and cracks of the former, when subject to strain/bending/deformation that are unavoidable for wearable, epidermal applications (see Section 1.2.2 for the details).

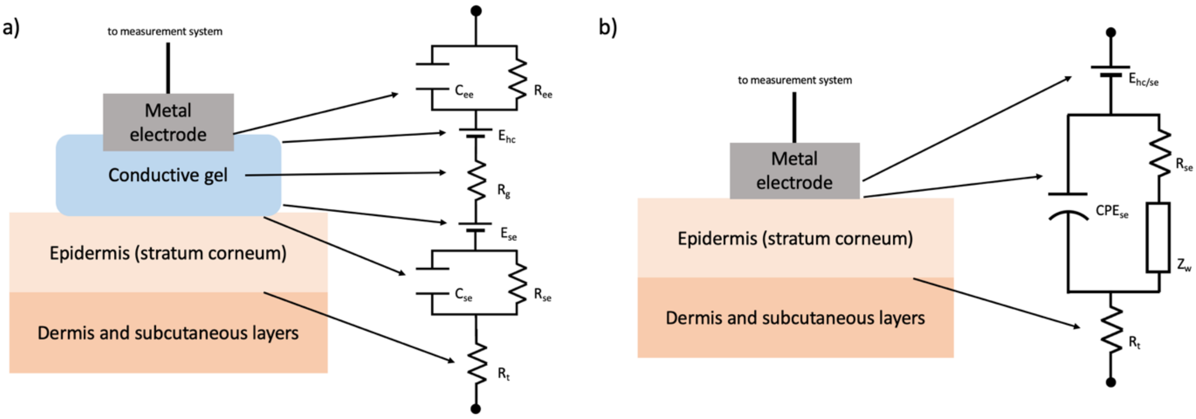


Fig. 1 : Schematization and electrical models of skin-electrode contact in the case of a) wet electrodes and b) dry electrodes.

Semi-dry electrodes and hydrogel based electrodes

Semi-dry electrodes were recently proposed as a trade-off between dry and wet electrodes, in order to overcome their limitations and issues. Semi-dry electrodes use significantly less conductive gel than wet electrodes (few tens of microliters vs. 1-2 millilitres) that is typically stored in an on-electrode reservoir, still retaining a low electrode/skin contact resistance. Typically, the electrodes are porous metal (such as porous titanium), to allow the electrolyte to permeate through the metal from the reservoir to the skin, thus providing a continuous wet interface between electrode and skin to maintain low and stable contact impedance and ionic/electronic current transduction. However, they are mechanically fragile and could be easily damaged, especially if employed in a daily life usage. If damaged, semi-dry electrodes could suffer from gel over-release (which risks bridging between different electrodes) due to on-electrode reservoir control issues. Moreover, in literature are reported some evidences of discomfort and skin irritation of the users after prolonged usage, due to the bulky structure of the on-electrode reservoir.

Very recently, a rising attention has been devoted to hydrogel-based electrodes: they are three-dimensional cross-linked polymer networks (typically an hydrophilic polymer), which can absorb and retain large amount of water. Hydrogels are the first biomaterials designed for use in human body and are finding widespread biomedical applications. Indeed, they offer great conformability and comfort for the patient, together with a tuneable flexibility. Indeed, they can be easily mechanically coupled with human epidermis and offer conformal contact with the skin. Moreover, thanks to their high water content and to the presence of conductive fillers, they show an extremely low impedance, especially when salts are added into the precursor of hydrogels during synthesis to improve the electrode conductivity. However, hydrogel based electrodes won't reach the mass and commercialization until there will be provided a full answer to several aspects: the biggest limitation of this approach is related to stability problems due to dehydration of the hydrogel via evaporation, which limits their life span to a few hours. Furthermore, the presence of salts can lead to skin irritation, and there are some concerns regarding a fully biocompatibility of these materials.

Epidermal/tattoo electrodes

Epidermal or tattoo electrodes refer to a class of electrodes with the ability to adhere and conform onto complex shaped surfaces, adapting their morphology to local details of the target surface (e.g. local roughness). This ability strictly relies onto both the intrinsic mechanical properties of the adopted materials and the structure/architecture developed to the electrode fabrication. In general, both the conformability and mechanical flexibility of a specific material are related to its bending stiffness, and they can be improved by reducing (i) the Young's modulus of the material and/or (ii) the total thickness of the material (see Section 1.2.2 for the details). Most of the reported approaches in literature for the fabrication of epidermal/tattoo electrodes are based on the second approach, i.e. on ultra-thin structures (with thicknesses ranging from few micrometres down to hundreds nanometres) transferred onto the skin by using a simple lamination process driven by van der Waal forces. Most of the interesting functional materials employed for epidermal electrodes are *conductive polymers* (CPs), such as *poly(3,4-ethylenedioxithiophene):polystyrene sulphonate* (PEDOT:PSS). Indeed, CPs offer features such as softness, biocompatibility and ultra-flexibility. PEDOT:PSS is the most known conductive polymer, wide employed in bioelectronics for its volumetric capacitance and mixed electronic/ionic conductivity, which makes them the ideal candidate to engineer the interface between the biotic (tissues) and the abiotic side (electronics components), due to the presence of the sweat that the skin continuously secrete. Therefore, CPs and in particular PEDOT:PSS, can transduce effectively the ionic currents into electronic currents, similarly to what happen for gel-based wet electrodes. The employment of conductive polymers, instead of metals, offer the possibility to dramatically reduce the level of thermal noise, power dissipation and signal degradation. Moreover, PEDOT:PSS offer an already established biocompatibility, which guarantee the opportunity of a prolonged on-skin use without any unwanted reaction, and an high robustness/mechanical resilience upon mechanical deformations. CPs can be deposited through cost-effective techniques (such as printing) making possible an easier upscaling of the fabrication at very low costs with respect to metals and nanomaterials counterparts.

1.2 SPECIFICATION OF THE EEG ELECTRODES

In order to quantify the actual capability of a specific electrode to be employed as transducer for EEG measurements, it is mandatory to define specifications, which must be measurable and verifiable. For the EEG devices, especially for prolonged monitoring, the electrical and mechanical stability as well as biocompatibility are considered as the most important figures of merit.

1.2.1 ELECTRICAL SPECIFICATIONS

As for the electrical specifications, the impedance plays a key role in maximizing the measured EEG signal quality. In this Section, we will attempt to give a rigorous definition the impedance in EEG systems. Practically, a EEG system (skin + electrode + recording instrumentation) can be electrically modelled with three types of impedance:

(i) the impedance of the whole skin;

(ii) the contact impedance between the skin and the electrode (at the ETI);

(iii) the impedance of the electrode;

As already reported in Section 1.1, the epidermis (or stratum corneum) is the most external layer of the skin and exhibits the biggest impedance, which is frequency dependent, and should be modelled with one capacitor and resistor in parallel. The value of the impedance ranges from 10 kΩ to 1 MΩ per square centimetre at 1 Hz, and mainly depends on the age of the subject and her/his hydration. The dermis and the subcutaneous layer are the internal part of the skin and it has been demonstrated that their impedance can be modelled by a single resistor (R_t, as reported in Fig. 1). The value for dermis + subcutaneous layer impedance is approximately 100 Ω per square centimetre, which is orders of magnitude lower than that of the epidermis layer, and therefore their contribution to the final impedance skin value is negligible.

The (iii) contribution – the impedance of the electrode – is mainly related to the material employed for the fabrication of the electrode, for instance metals (gold is the most employed option), conductive polymers, nanomaterials etc. The impedance value depends also on the shape and the area of the electrode, and other parameters (impurities, interlayers etc.). In general, the (iii) contribution is several order of magnitude lower of the epidermis impedance, and does not play any role on the final impedance value.

The (ii) contribution – the impedance between the skin and electrode – plays the main and key role in determine the impedance in EEG systems, since it has the highest impedance value of the path skin-recording instrumentation. Again, its value can be finely tuned by modifying the architecture/size of the electrode and/or the functional material employed as electrode.

Overall, assuming (i) Z_s the impedance of the whole skin, (ii) Z_{se} the contact impedance at the ETI and (iii) Z_e the impedance of the electrode, we can calculate the impedance of the EEG system Z_{TOT} as the series of the previous three contribution, as follow:

$$Z_{TOT} = Z_s + Z_{se} + Z_e \cong Z_{se} \text{ (if } \mathrm{Z_s} \ll \mathrm{Z_{se}}), \qquad (1)$$

When Z_{se} is very high (> 100 MΩ), it behave like an antenna and absorb environmental noise, especially at 50/60 Hz. Fortunately, Z_{se} (and consequently Z_{TOT}) is typically well below this critical value. Specifically, for wet electrodes the contact impedance at the ETI is in line with that of the whole skin (i.e. epidermis), thanks of the

use of a conductive paste/gel or saline solution, which increase the contact area and favour the ionic/electronic current transduction, as widely reported in Section 1.1.
As for the dry electrodes, the Z_{se} typically exceeds the values from hundreds of kΩ to tens of MΩ, while in the case of epidermal electrodes and hydrogel based electrodes it has been reported values of contact impedance of few tens of kΩ, in line with wet electrodes. In addition, epidermal electrodes based on ion-permeable can further to reduce the skin-electrode impedance when they are in contact to the sweat that is continuously secreted by the skin, thanks to their volumetric capacitance. This aspect, together with a well-established biocompatibility (not achievable with nanomaterials or hydrogels), makes the CP based electrodes one of the most appealing option as electrode for EEG recording.
Quantitatively speaking, although an impedance Z_{se} of 10-100 kΩ can be considered low enough for the recording of several biopotentials through skin electrodes (such as electrooculography, electrocardiography, electromyography), this is not the case as for EEG signals. Indeed, since EEG signals are typically small – in the range from few to tens of µV due to their propagation through the head volume and to the weak reliability of electrode-skin connections – the engineering of the ETI must be carefully optimized in order to reduce Z_{se} to values below 10 kΩ. Clearly, large area electrodes can be used to reduce the Z_{se}, although this approach would lead to a significant reduction of the spatial resolution of the measurement. The typical methods to measure the skin-electrode contact impedance are the 3-terminal and 2-terminal measurements reported in Fig. 2. These methods are well-known and their description is beyond the scope of this document.

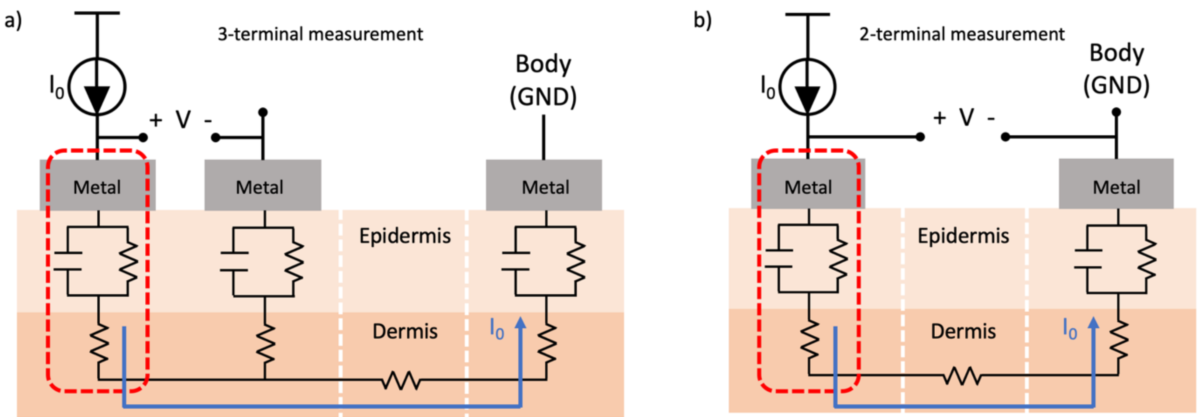


Fig. 2: How to quantify skin-electrode contact impedance. a) 3-terminal method and b) 2-terminal method.

1.2.2 MECHANICAL SPECIFICATIONS

Besides the electrical specifications, another important aspect that need to be considered and tackled is the mechanical stability and robustness of electrodes-skin contact. On one hand, a stable and robust electrode-skin contact leads to a stable contact impedance. On the other hand, mechanically robust electrodes can mimic the mechanical properties of the skin and, therefore, sustain stresses (such as bending, elongation, compression etc.) that are unavoidable for on-skin, daily-life applications.

Indeed, the human skin has the capability of undergoing deformation under the influence of external forces, such as the weight. Similarly to elastic materials, it is subject to the mechanical laws that define its properties. As a matter of fact, in the so-called *elastic range* the skin can be stretched to several times its original size, maintaining its original mechanical properties and returning to the initial state when the external force is removed.

The relation between the *mechanical stress* (σ) and *strain* (ε) is linear in the elastic range, and is described by the *Hooke's Law* that characterizes the resistance to elastic elongation:

$$\varepsilon = \frac{E}{\sigma}, \quad (2)$$

where the factor of proportionality (E) is the *Young modulus* (i.e. the modulus of the longitudinal elasticity). Some interesting works in literature reported that the Young modulus of the skin varies between 420 kPa and 850 kPa for torsion tests, 4.6 MPa and 20 MPa for tensile tests and 50 kPa and 150 kPa in the suction tests. These large ranges depend on the condition of the skin (mainly hydration) and the age of the subject. Indeed, as all of us may have experience, in the process of ageing the skin becomes stiffer, since its Young modulus increase, less elastic and less flexible. Regarding the actual stretchability of the skin, it has been reported that it can reach the impressive value of 70% strain and, consequently, has the capability to be adapted to the movements of human's body by bending and stretching. Again, this value is strongly related to the age of the subject and the condition of the skin.

Overall, in the development of electronic systems that are fully integrable onto the skin, the materials chosen for their fabrication should reflect the flexibility and stretchability of natural human skin. Therefore, the previous values should be considered as a guideline.

As widely reported in the previous Sections, wet and semi-dry electrodes offer a mechanically stable interface between the electrode and the skin: indeed, the Young Modulus of gels typically is below 100 kPa and they are soft and capable to conform to any complex 3D, rough surface. However, they are not practical for daily-life applications because the mechanical stability of the contact is ensured only for few hours – i.e. before the drying of the gel.

Hydrogel based electrodes have a Young Modulus that is typically in the order of few tens of kPa, and they offer great conformability and comfort for the patient, together with an extremely high flexibility. Again, the biggest limitation of this approach is related to stability problems due to dehydration of the hydrogel via evaporation, which limits their life span to a few hours, and biocompatibility problems (skin irritation).

Dry electrodes suffer of instability since the forces applied to put in contact the electrodes and the skin (such as headset) are typically not reproducible and not comfortable at all. Therefore the measurement of weak biopotentials, such as EEG, results into a too high variability, especially in terms of frequency response.

For these reasons, among all the different solutions presented in Section 1.1, epidermal/tattoo electrodes are probably the best option for long-term, daily-life applications. Indeed, they guarantee a conformal and imperceptible adhesion with human skin, achieved by a simple lamination process driven by Wan der Walls forces, without any additional adhesive or conductive paste. Moreover, they ensure an extremely high mechanical resilience and stability, thanks to their ultra-low thickness (from few hundreds of nm to few tens of µm). Indeed, the reduction of the thickness of flexible materials is crucial to achieve better conformability, mechanical stability of the contact and resilience, due to the reduction of the *bending stiffness* (EI) of the structure, which determines how flexible and conformal (to the underlying substrate) a material is.

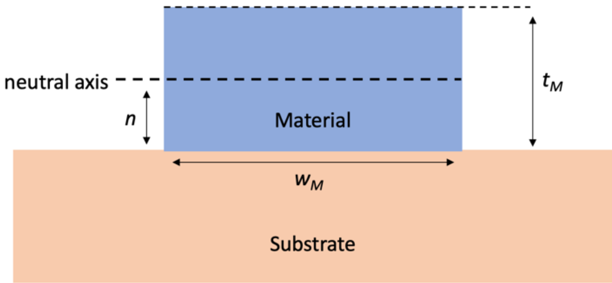


Fig. 3: Thin-film material laminated onto a substrate. The thin-film bending stiffness depends on its Young Modulus (E_M) total thickness (t_M), width (w_M), and the distance of its neutral axis to the substrate (n).

The bending stiffness of a material depends, with a good approximation, on its Young Modulus and has a cubic dependence on the material thickness. This relationship is quantified by the following equation:

$$EI = E_M w_M t_M \left(\frac{1}{3}t_M^2 + t_M n + n^2\right), \qquad (3)$$

where E_M represent the Young modulus of the material – laminated onto a generic substrate – while w_M, t_M, and n represent its width, thickness, and distance to the natural axis, as reported in Fig. 3. Therefore, the bending stiffness can be minimized by reducing the thickness of the material or its Young Modulus. Since the tunability/variation of the Young Modulus of materials is rather complex, most of the reported approaches in the literature for the fabrication of ultraflexible and ultraconformable electrodes are based on super-thin (few micrometers thick) or ultra-thin (thickness below 1 µm) non-stretchable materials (Young modulus in the order of GPa). For these structures, the physical flexibility and stretchability can be easily achieved by laminating these thin-film electrodes onto a pre-stretched, soft substrate, such as an elastomer (*film-on-elastomer* or FOE) or human skin. Upon the relaxation of the substrate, the stress at the electrode-substrate interface, originating from the mismatch between the Young Moduli of the substrate (typically hundreds of kPa) and the electrode, induces wrinkles onto the thin-film, as depicted in Fig. 4. Subsequently, the thin-film can be repeatedly stretched up to the pre-lamination limit. Epidermal electrodes benefit also from another important advantage, related to the maximum strain that they can sustain before the damaging of the substrate (typically cracks onto the surface). Indeed, ε depends on the bending radius (i.e. the radius of the circle that fits the curvature of a bent material) and the thickness of the bent material. In the case of a epidermal electrode, the induced strain can be described by the following equation:

$$\varepsilon = \frac{d_M}{(2 * R)}, \quad (4)$$

where d_M is the thickness of the thin-film and R is the bending radius. Therefore, the lower is the thickness of the electrode, the lower is the induced strain, for a fixed bending radius.

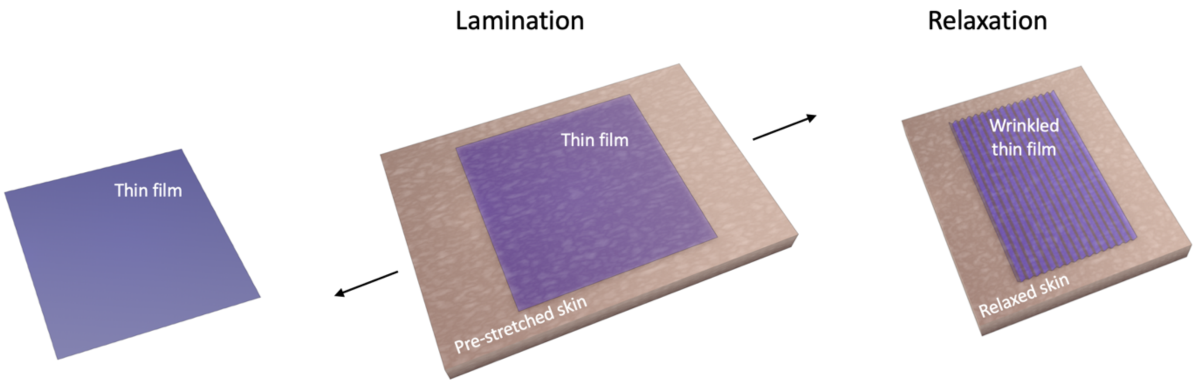


Fig. 4 : Schematic procedure to induce wrinkling in thin-films.

Mechanical properties of thin-films: how to measure it?

The already mentioned film-on-elastomer (FOE) system is a well-known approach to give an "apparent" stretchability to non-stretchable thin-film laminated onto a soft elongated elastomer. In addition, through FOE is possible to measure the mechanical properties of thin-film with sub-micron thickness. For instance, properties such as the Young Modulus, the *yield point* (Y) and the failure point or *crack-onset strain* (CoS) can be easily measured for FOE systems. The yield point is the strain at which an elastic material under increasing stress ceases to behave elastically; under conditions of tensile strength the elongation is no longer proportional to the increase in stress and the material begins to plastically or permanently deform. While the crack-onset strain (CoS) is a measure of material ductility and is a significant parameter to consider when designing devices. Indeed, it gives information on the maximum strain at which a film fractures.

A stress-strain curve for a hypothetical polymeric thin-film is shown in Fig. 5.

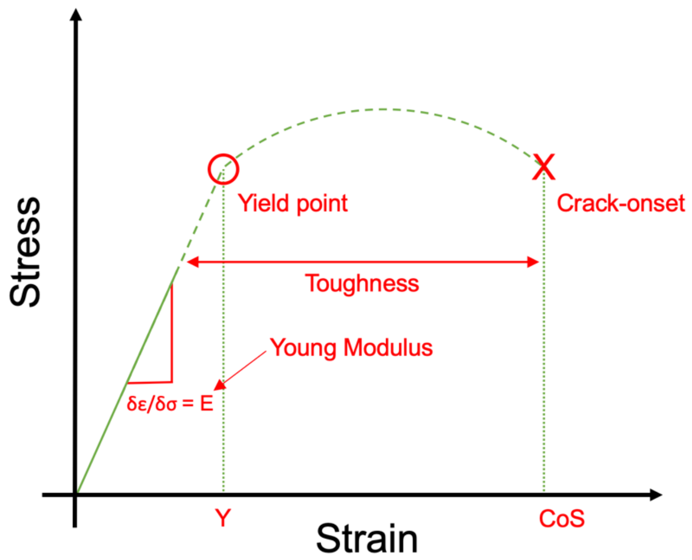


Fig. 5 : Stress-strain curve for a specific material.

The Young Modulus can be measured with the *buckling-based metrology* for FOE systems: the material of interest is transferred onto an elastomeric substrate that has been pre-stretched by a few percent. Upon the relaxation of the elastomer, the film then forms sinusoidal wrinkles out of plane. The distance between the peaks of the wrinkles (i.e. the wavelength of the wrinkles) is defined by the energy balance between the energy required to deform the substrate and the energy required to deform the thin-film. The Young Modulus of the film (E_{film}) can be extracted from the following equation:

$$E_{film} = 3E_s\left(\frac{1-\nu_f^2}{1-\nu_s^2}\right)\left(\frac{\lambda_b}{2\pi d_f}\right)^3, \quad (4)$$

where λ_b is the buckling wavelength, d_f the film thickness, E_s the Young Modulus of the substrate, ν_f and ν_s the Poisson's ratios of the thin-film and the substrate, respectively.

The CoS can be determined by observing the formation of cracks, for instance by atomic force microscopy, in a thin film on elastomer under increasing strain. The yield point of a material can be measured by using the diffraction of light from a laser (LADYP). With this method, a thin-film is laminated onto a relaxed elastomer and then the system is subject to cycles of tensile strain that gradually increase in step of 1% and then released (i.e. 0% → 1% → 0% → 2% → 0% → 3% → 0% → 4% → 0% etc.). The formation of wrinkles manifests as a diffraction pattern obtained using a laser, and represent at first the yield point of the polymer and then the CoS.

Besides FOE based methods, there are other techniques to measure mechanical properties of thin films. The Young Modulus is commonly measured by *nanoindentation*, in which a thin-film is indented with a cantilever tip. However, if the thickness of the film is below few micrometres, the (E) measurement with technique is not trivial.

1.3 OBJECTIVES AND WORKFLOW OF WEAR PROJECT

As already introduced in the Abstract Section, the wEAR prototype is an innovative lab-based hearable (i.e. wearable on-ears) technology that can estimate and decode the wearer's auditory attention, by processing her/his electroencephalography (EEG) signals with enhanced sensing capabilities. The core of the wearable on-ear system for the recording/processing of EEG signals are the functional thin-film electrodes their interface with the skin.

The conductive polymer poly(3,4-ethylenedioxithiophene):polystyrene sulphonate (PEDOT:PSS) blended with an ionic liquids will be initially tested as conductive material. It has been reported that by the addition of a small amount of ionic liquid to PEDOT:PSS is beneficial for the reduction of its impedance (electrode conductivity and skin-electrode contact impedance) as well as an improvement of the mechanical properties of the polymer. We will test different formulations and optimize the deposition techniques, towards reaching the goal of low impedance electrodes, with a thickness in the range of tens of µm. Several figures of merit will be taken into account for the validation of the layout and structure of the electrodes: firstly the electrical and the skin-electrode contact impedance, secondly the mechanical resilience (strain and bending test) as well as conformability (adhesion test). The measured values will be then compare with the specification reported in this Deliverable, which will be, therefore, the guideline for the fabrication of the electrodes, during the last four months of WP1 (M3-M6), and their electrical/mechanical characterization, during WP2 (M3-M6).

After the validation of the conductive polymer based thin electrodes, their use for recording the EEG signals from the ears of normal-hearing volunteers in both "single sound source" situations during WP3 (M7-M11), and "cocktail party" situations during WP4 (M10-M15). In parallel, ear-EEG forward models, which are a powerful tool for mapping electrical sources in the brain to potentials recordable from the ear (inner or outer), will help to define crucial aspects, such as number of electrodes and their location (design), reference electrode (which can maximize the signal-to-noise ratio). Initially, well understood and characterized signals will be measured, including auditory evoked potentials (auditory brainstem response, frequency following response, cortical responses), and brain rhythms (alpha, beta, theta) by evaluating comparators such as: amplitude of the signals, time resolution, noise, robustness and stability of the skin/electrodes connections, amplitude at the ear, and long-term reliability.

In WP5 (M7-M24) we will develop the tools/methods for the estimation of the attended sound source, as well as the whole data analysis that will last until the end of the project. In particular, for the extraction of the attended sound source, using only signals available in the vicinity of the ear, we will use linear models that map EEG temporal responses to sound signal fluctuations, the so-called temporal response functions (TRFs). For wEAR project, we will be particularly interested on "backward" TRF models, which map brain response to speech envelope following an envelope reconstruction approach. Machine learning algorithms will be surely of great support for the real-time implementation of such cognitive steer of the wEAR device, for instance vector machine classifier models. During this phase, it will be imperative to have a particular attention to the potential portability of the whole processing chain as this would have to eventually run on a compact and low-power processing unit. This unit will be a self-contained module (based on standard electronic components – amplifier, filters, battery etc.) for the EEG data processing: pre-processing (artifacts/noise removal), features extraction and classification. Particular attention will be devoted to the integration of the module with the thin functional electrodes, in order to reduce the high stress generated at the interface: it will be followed a magnetically interfaceable thin-film approach, using an intermediate interconnection layer made up of a composite thin film based on the combination of the conductive polymer PEDOT:PSS and ferrimagnetic powder. The methodology which will be followed for this project is reported in Fig. 6.

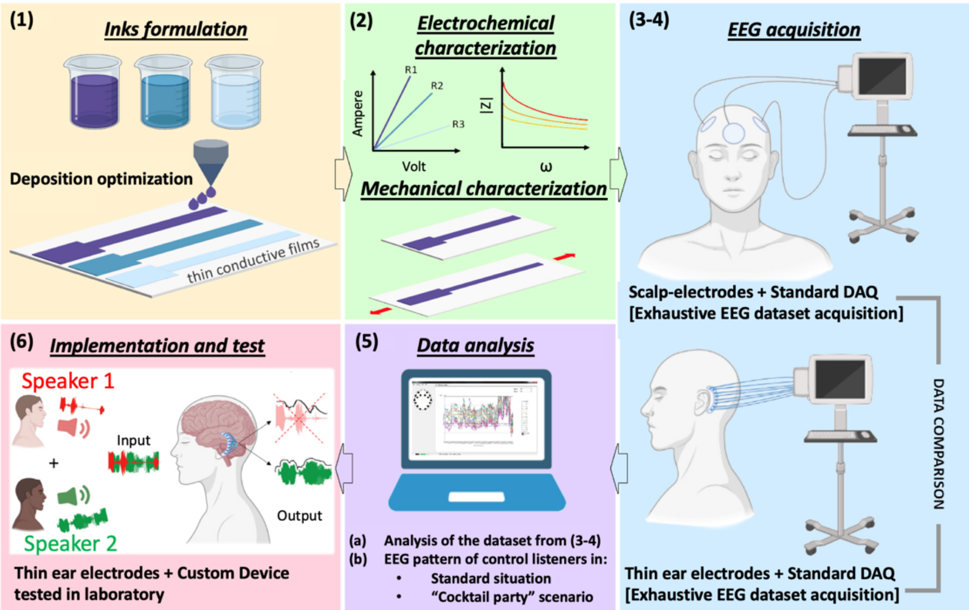


Fig. 6 : Research approach and methodology.

CONCLUSIONS

Deliverable 1.1 details the main electrical and mechanical specifications of the wEAR electrodes. The overview pf the project and the objectives/goals to be realized in each WP are also included. All the specifications have been defined to achieve a portable, cost-effective device.

While the wEAR platform is in principle a widely applicable technology, since it could be employed to record any kind of biopotentials, this project targets in particular the recording of EEG signals from the ear of volunteers. The system specifications here provided have been defined after a careful analysis of the state of the art.

The wEAR platform will be fabricated making a substantial use of novel cost-effective mass-manufacturable, large-area compatible, scalable techniques such as printing (screen or ink-jet) or coating (spray coating). The system includes *ad-hoc* designed electronics for the acquisition and processing of the EEG signals.

The project final aim is the discrimination of the Auditory Attention of the user. This main focus is enabled by the unprecedented sensitivity of the wEAR electrodes, together with an high level signal processing and data analysis. As already mentioned, the wEAR platform could then be used for many other applications in the field of precision medicine.

This document will be updated every six months, if needed, to check the relevance of the specifications, their match to clinical needs and to adjust them whenever necessary.