# Comparison of Standard and Microneedle Array Electrode Impedance of Skin Electrode Interface for EEG signals: A Review*

Felipe A. Sulez[1], Martha L. Zequera[2], and Manuel R. Perez[3]

Pontificia Universidad Javeriana, Bogotá D.C, Colombia
sulez.fa@javeriana.edu.com
mzequera@javeriana.edu.com
manuel.perez@javeriana.edu.com

**Abstract.** This paper presents a literature review of dry electrodes based on microneedle technology to measure biological signal and its advantage in regard with wet standard electrodes. Surface electrodes require skin preparation and conductive gel to maintain low interface impedance, which limit biomedical applications in long-term monitoring of the electroencephalogram (EEG) signals. The skin electrode interface plays a very important role in the acquisition of high-quality biological signals due to its low impedance requirement in order to reduce the transduction resistance among ionic and electrical currents. Wet electrodes are currently using, which require abrasive/conductive gels to improve the conductivity of the skin. Advances in dry electrodes, which do not require gels, have simplified this process and contributed to maintain a good electrical contact with the skin surface. The objective of this paper is twofold: i) to provide an overview of electrodes using currently based in microneedle array and ii) to compare it with traditional wet-based electrodes used to acquire of EEG signals for medical applications.

**Keywords:** Skin Electrode Interface · Dry Electrodes · Wet Electrodes · Conductive Gel · EEG Signals.

## 1 Introduction

Bioelectrodes play an important role in medical and biomedical applications. They are designed for measuring and recording biological signals and have the ability to transduce the bioelectric activity in the body (ionic current) into electrical current that can be measured and recorded [1]. The electroencephalogram (EEG) records the electrical impulses of the brain and is an essential tool for monitoring in real time the brain health conditions and the detection of abnormal neurological activity, which contains rich medical information about brain activity. EEG signals are very small in amplitude ($\pm 100\mu$V) and are susceptible to many types of noise interference from skin, electrodes and saturated amplifier

* Supported by organization x.

signals [2]. Recent advances in electronics amplifiers has lead to the reduction of its noise, as a result they are not anymore the dominant noise source in biopotential recording. In contrast, electrodes introduce noise components and the performance of non-invasive electrodes in detection of biological signals are highly dependent on skin electrode (SE) interface impedance. Therefore, the low electrode-skin contact impedance is a fundamental requirement to obtain a good quality signal in order to reduce the transduction resistance among ionic and electrical currents, which occurred at the metal area of the bioelectrode. A good impedance behaviour and modelling allows high resolution in surface biopotential recording.

Therefore, to record accurate EEG signals, a skin preparation procedure is required, which involves the abrasion of the top layer of the skin, the stratum corneum (SC) and the application of conductive gels to improve the electrical conductivity. The EEG monitoring process begins by locating the sites for electrode placement based on the international 10–20 system [3]. This removes the SC, which is the largest contributor to the SE interface impedance. The SC impedance is inversely proportional to frequency, ranging from 200 k$\Omega$–200$\Omega$ over a frequency range of 1Hz–1MHz [37]. The most common electrodes used for measurement EEG signal are flat of cup shaped metal of Silver-Silver Chloride (Ag/AgCl) electrodes. They must be coated with conductive gel to improve the conductivity in the skin and reduce the impedance present in skin electrode contact. Conductive gel typically contain chloride ions ($Cl^-$), which play an important role in establishing a non-polarized electrode-electrolyte interface and stable SE interface [32].

The Ag/AgCl electrodes are the most popular non invasive electrode in clinical applications, considerate as a gold standard to measure small amplitude biological signal [34], which have been widely studied and used as non-invasive low-impedance SE for providing high quality signal recording. Ag/AgCl electrodes also have some limitations, such as a skin preparation time and discomfort. Another disadvantage is the use of the conductive gel for a long time (1-2 hours) since it tends to dry out and it is easily influenced by skin conditions, such as sweat accumulation, degrading the EEG signal quality [4].

To avoid the aforementioned drawbacks of standard wet electrodes, various types of dry electrodes have been introduced and discussed in various publications [27], [38], [39], [40]. There are two main categories, contact and non-contact dry electrodes. The non-contact electrodes are capacitively coupled to the skin, hence these electrodes are appropriate for measurement of small amplitude of biological signal as EEG but are highly sensitive to movements and increase of the SE impedance. The contact electrodes are based on microneedle array (MNA) that penetrates the SC skin layer and makes directly contact with the living inner skin layer, being a promising technology for measurement of EEG signals. In fact, a dry electrode is not absolutely dry because a tiny amount of sweat and moisture is always present. Undoubtedly, the dry MNA electrodes have shown many advantages compared with Ag/AgCl electrode such as rapid preparation set up of the skin, comfort, low impedance of the skin, long-term

monitoring, minimal skin trauma, painless puncture without skin preparation, reduced infection, high selectivity, easy post record cleaning and elimination of conductive gels than in most of the cases can result in skin irritation and damage [34], [36]. As mentioned before, a standard electrode caused some hard wounds in the patient due to the use of conductive gel and therefore is not suitable for long-term monitoring. The MNA electrode without explicit conductive gel reduce the high impedance of the stratum corneum (SC) layer and the undesired electrode motion. The surface microstructure array of a MNA can increase the actual contact area between the electrode and the skin at the same contact surface, resulting in the decreasing of the (SE) impedance. In the MNA electrode maintain a good contact status with the skin when a relative sliding occurs at (SE) impedance interface, which avoid the electrode motion and contribute to have a higher signal-to-noise-ratio (SNR) [34]. For these reasons is considerate an alternative to measurement EEG signal. This paper is organized as follows: Section 2 presents the Methodology and methods used for the comparison of the two technologies. Section 3 describes the electrodes models employed for the analysis and the comparison among both technologies. Section 4 presents the the analysis of the structural variation of electrode parameters. Section 5 shows a brief discussion about. Finally Section 6 gives the remarks and the conclusions of this review.

## 2 Methodology and methods

According to Huigen et al, the SE impedance plays a very important role in the acquire of biological signal quality and high SE impedance influences negatively in biological signal quality since it is associated with low SNR [42]. In the other hand T. McAdmams et al, the MNA electrodes are designed to eliminate the need for electrolytic gels, which makes the installation process simple with a short setup time and also it prevents an increase of the impedance due to drying of gels [41]. However, the absence of conductive gels means that controlling the contact impedance at the SE interface is more difficult than using the conventional wet electrodes. Another considerations of the literature review were taking into account such as the impedance characteristics and the surface contact [5]. In this sense, several researches has been implemented MNA technology in their researches on medical and clinical applications.

In this paper the authors consider four essential parameters that may produce impedance variation as follows: i) distance among electrodes, ii) contact SE interface, iii) material properties and iv) size and geometrical design. Based on these parameters a deeper literature review was made on sate of art of MNA electrodes (Fig. 1) in order to may compare traditional wet-based electrodes with MNA electrodes used to acquire of EEG signals for medical applications.

A systematic search review was made by 42 articles and mainly in the last 9 years (2010-2019) will describe the progress in micro mechanical system of MNA according to its relevance in the biomedical and clinical environment. The research line show the main historical aspects during of development of

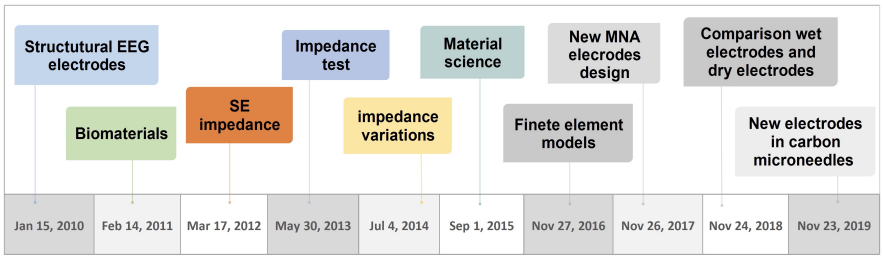


**Fig. 1.** Research line of state of art of MNA electrodes (2010-2019).

the MNA technology. According with this research line, since 2010 begins the approaches of structural EEG electrodes to allow a completive impedance in the fabrications and manufacturing process. New challenges were reported in range of period (2016-2019) of MNA electrodes based on emerging biomaterials, innovative designs and new parameters to improve SE interface impedance.

## 3 Impedance and MNA electrodes

As it was mentioned previously, the impact of impedance in quality of biological EEG signal is very important to provide adequate biological signals according with the standard records. In this section we expose the relation of the four elements previously mentioned in the methodology section, that have a highly impact in the variations of impedance.

Depending of the method to collect the EEG signal, there are two acquisition types, invasive and non-invasive. The invasive method requires the surgically implanting electrodes on brain surface, or even on the same brain depth. Its use is commonly extended to patients who due to their clinical condition are desired get each and every brain signal with good quality. In a non-invasive procedure the location of the electrodes on the cranial surface is according by the international system 10-20; standardized by the American Electroencephalography Society (AES) and recommended by the International Federation of Electroencephalography Associations and Clinical Neurophysiology [10].

A data acquisition system for EEG signals provides a consistent and applicable method of recording EEG with 21 electrodes placed at relative distances (10% or 20%) between the cranial landmarks over the head [6]. It has also been used as a relative standard head surface–based positioning method for recording evoked and event-related biopotentials and for various transcranial brain mapping method. To display the distribution of EEG activity, it is necessary to use Ag/AgCl electrodes commonly named in the literature as wet standard electrodes (Fig. 2). Those electrodes should not significantly attenuate signals between 0.5 and 70 Hz [7]. Experimental test suggests that silver-silver chloride or gold disk electrodes held on by collodion are the best, nevertheless electrode

materials and electrode conductive gels have been used effectively, especially with amplifiers having high input impedances [11]. The Ag/AgCl electrodes require abrasion of conductive gel to increase the electrical conductivity on skin and decrease bio-impedance, a condition that varies from person to person and it cannot be controlled [8]. The bio-impedance decrease in a frequency range of 100 Hz to 2 Khz applied a density current between 0.25 to 1000 $Am^{-2}$ [12].

Electrolytic gel in combination with a foamb backing help to fix the electrode position. However, the electrochemical reactions that take place on the contact surface produce fluctuations in the bio-potential that can cause an increase of noise levels up to 10$\mu$V peak-to-peak [13]. The conventional wet electrode requires skin preparation such as hair-cutting, skin abrasion and electrolytic gel coating, which are time-consuming. The usage of conductive gel is likely to cause skin irritations, allergic reactions, and even skin damage in neonates [9]. Ag/AgCl electrodes are no suitable for long-term bio-signal recording, owing to the gradual drying of conductive gel and increase in the (SE) interface impedance and the resultant reduction in the quality of collected signals.

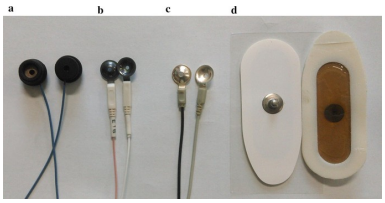


**Fig. 2.** The four most used wet standard electrodes for EEG.

In general, comparative bioimpedance studies performed in Ag/AgCl electrodes have reported high impedance, approximately 10e7 Ohms [14], [15], [16], [17], by the addition of conductive gel on skin which contributes to improve the conductivity on skin.

Thanks to new advances in materials and electronic systems technologies, a new generation of dry electrodes has been developed to fulfill the need, in order to record quality EEG signal without the application of electrolytic gel. The MNA should be designed to penetrate through the stratum corneum and epidermis layer of human skin. This enables it to bypass the stratum corneum, which has high impedance characteristics, and it establishes a direct contact with the out layer skin. Moreover, minimize the trauma to achieve painless monitoring, the microneedle tips should not reach the dermis to stimulate the pain receptors and puncture the blood vessels. In the scientific literature, there are different approaches of MNA technology about biomaterials, structure, surface contact, types of arrays, and geometrical sizes.

According with the biomaterials, these are essential for the sensitivity and selectivity of an impedance measurement system. The selection of material de-

pends of the purpose of use, the inertia of the material to the environment, the complexity of the manufacturing and the cost [22]. In order to simply the manufacturing process, in general only one material is adopted for an electrode in applications of electrophysiological signal measurement. The electrode materials commonly used are Ag/AgCl, gold (Au), metallic nanomaterials, and carbon based nanomaterials [31]. In the same way, the authors in [29] proposed new dry electrodes based on flexible materials as an alternative to replace the Ag/AgCl electrodes. Electrodes based on flexible materials (e.g., polymers), tackle these limitations as they ensure better use and comfort for the patient, since the electrode surface aligns to the curvature of the body shape and therefore it reduces the motion during long-term bio-signal [18]. Polymers such as polysiloxane [19],[20], polyurethane [21], and parylene [23] have previously been used as substrate materials.

Another alternative to MNA electrodes is developed by the authors in [24] where they proposed a electrode based on three types of microneedle arrays (6x6, 7x7 and 9x9 array, respectively) with a hight of 500$\mu$m, 400$\mu$m and 300$\mu$m, respectively located over an area of (5mm x 5mm). To penetrate the skin easier, the tip of the microneedle takes a pyramidal cut shape. At the same time, its length is also an important parameter for drilling the germinative layer and stop at the dermis layer. Because the average thickness of the epidermis is approximately hundreds microns, the thickness of the stratum corneum layers and the germinative layer is approximately 10-15$\mu$m and 50-100$\mu$m respectively [33]. The total height of the microneedle is 200$\mu$m and the tip should not exceed 20$\mu$m [24]. Regarding the impedance results, this is directly proportional to the increase of contact area. The Figure 3 shows a typical microneedle electrode.

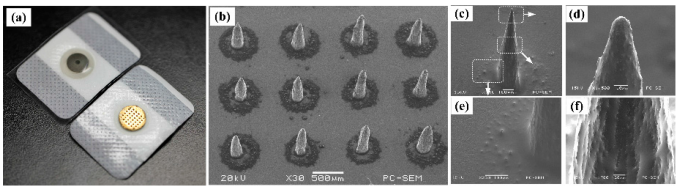


**Fig. 3.** Characteristics of MNA elecrodes for EEG [24]

In addition, Wang et al developed a MNA based on flexible material using parylene. The height of microneedles should reach up to 100 $\mu$m to minimize the pain resulting from penetration, the impedance test is comparable to that of wet electrode without the skin preparation in the frequency range of 10–40 Hz [25]. In the same way, Ren et al describe a curved electrode which consist of a 5x7 array whose geometry is conical located on the conductive surface. The microneedles were covered with a Ti/Au film, which allows to improve the adhesive strenght

of the microneedle array, as well as avoiding a direct interaction with the tissue cutaneous guaranteeing the biocompatibility of the sensor [26].

Regarding to the SE impedance variation in frequency, Gosth et al and Anand et al show a Finite Element Model (FEM) of the behaviour of impedance and the distance among electrodes analyses and they reported that the distance is directly proportional to impedance but not to the electric field [30],[31]. Although, there are other electrode alternatives to acquire EEG signal such MNA based on carbon materials that maintain a relatively good contact on skin and competitive impedance [27], [28].

## 4 Swot Analysis between Wet Standard and MNA Electrodes

The current study made on electrodes was based on a swot analysis between wet and MNA electrodes in order to established, strengths, weaknesses, opportunities, and threats will describe the main advantages and disadvantages of both technologies used to performance the impedance at SE interface impedance in biomedical applications. (Table 1)

## 5 Discussion

While most EEG monitoring uses wet standard electrodes that require some scalp/hair preparation procedures in order to make contact between the electrodes and bare scalp, in this study, we investigated the feasibility of EEG measurement without use of conductive gel.

Using the wet electrodes in previous studies[14], [15], [16] we found that the modification of electrode surface deteriorate its performance and the quality of the EEG signal, in terms of the SNR. In addition, surface Ag/AgCl electrodes are the most common and favoured electrodes in clinical measuring for recording small amplitude bio-signal (e.g EEG, ECG, EMG) with a relatively high resolution but they are associated with high electrode skin impedance and motion artefacts.

MNA electrodes can successfully penetrate through the stratum corneum into skin, reduce the electrode-skin interface impedance, and achieve accurate biological signal recording without skin preparation or coated of conductive gel. The MNA exhibits similar or even higher bio-signal recording performance in comparison with the conventional wet electrode. The MNA is a promising electrode for long-term bio-signal recording. The wet electrode is not suitable for long-term bio-signal recording an require the use of electrolytic gel. The use of conductive gel change the properties of skin and dry out in one to two hours and deteriorate the quality of bio-signal. However, certain limitations of the MNA in clinical environment still remain. The safety of MNA requires further investigation, particularly, the biocompatibility of the materials selected for the manufacture process.

| Strengths, Weaknesses, Opportunities, and Threats Analysis | | |
|---|---|---|
| | Wet Standard Electrode | MNA Electrode |
| Strengths | They are the most common electrdoes in clinical measurement for recording small amplitude biosignal [34], [36] | A high SNR is introduced by the direct contact of micronnedle with the skin |
| | Wet electrodes are non invasive | The MNA electrode resulted in lower impedance than the commercial electrode, despite the absence of skin preparation and electrolytic gel application [19] |
| | Can be maintain a relatively good contact on skin | It can penetrate through the stratum corneum and eliminate the influence of stratum corneum on the impedance [32] |
| | They are low cost fitting with skin and nice condcutivity | They can obtain bio-signals with a realatively high resolution and provide stable data quality over long recording sessions |
| Weaknesses | They are not suitable for long-term biosignal recording owing to the gradual drying of electrolytic gel and inrease in the skin interface impedance [41] | They are classificated as invasive electrode by makes directly contact with the inner skin layer |
| | Requiere skin prepreparation such as hair cutting skin abrasion electrolyte gel coating [35] | MNA electrodes are succeptible to motion artefacts due to the lack of an electrolytic layer |
| | The use of electrolytic gel is likely to casue skin irritation allergic reactions, and even skin damage in neonates [9] | |
| | They are succeptible to motion artefacts | |
| Opportunities | Improve electric properties of conductive gels coated in the surface of the contact area | New designs of the microneedle, biomaterials and competitive impedance in SE interface. |
| Threats | Skin overhydration would greatly change skin impedance | The penetration process is almost painless, as its small size prevents the stimulation of dermal nerves and evading the generation of the pain sensation |
| | The conductive gel dries out, failing to meet the stability of signal quality requirement | |
| | Wet electrodes cannot reach the conductive layers of the epidermis due to the stratum corneum which hinders the extraction of signals | |

**Table 1.** Technical parameters used in wet standard MNA electrodes reported by the most relevant studies during (2010-2019).

## 6 Conclusions and outlook

Recent significant progress in MNA electrodes has enabled many applications for out-of-hospital EEG monitoring in the areas of healthcare and Brain Computer Interface (BCI). The MNA electrode can collect an EEG signal detected that compared well with the one recorded with a standard wet electrode. In addition, its performance is better than that of the conventional rigid-surfaced capacitive electrode. Future work should focus more on the fabrication of MNA with high mechanical strength and adequate biocompatibility. Most of the MNAs were fabricated on rigid substrates that are not conformal to curved and moved human skin.

## References

1. Walls-Esquivel E., Vecchierini, M. F., Héberlé, C., and Wallois, F., "Electroencephalography (EEG) recording techniques and artefact detection in early premature babies," Neurophysiologie Clinique/Clinical Neurophysiology, vol. 37, 2007, pp.299-309.
2. Blinowska K., and Durka P., " Electroencephalography (eeg)," Wiley encyclopedia of biomedical engineering, 2006.
3. M. Teplan, "Fundamentals of EEG measurement", Measurement science review, vol. 2, 2002, pp.1–11.
4. H. Hinrichs, M. Scholz, A. Baum, J. Kam, R. Knight, and H. Heinze,"Comparison between a wireless dry electrode EEG system with a conventional wired wet electrode EEG system for clinical applications. Scientific Reports," Scientific Reports, vol.10, 2020, pp.1–14.
5. S. Lee, Y. Shin, A. Kumar, K. Kim, H. Lee, "Two Wired Active Spring-Loaded Dry Electrodes for EEG Measurement," Sensors, vol 19., 2019, pp.4572.
6. J. Acharya, A. Hani, J. Cheek, P. Thirumula, and T. Tsuchida, "American Clinical Neurophysiology Society guideline 2: guidelines for standard electrode position nomenclature," the Neurodiagnostic Journal, vol 54, 2016, pp.254–252.
7. American Clinical Neurophysiology Society. "Guideline 1: Minimum technical requirements for performing clinical electroencephalography," Journal of clinical neurophysiology: official publication of the American Electroencephalographic Society, vol 23, 2006, pp.86.
8. S. Hyem, and M. Attari, "Electrode-gel-skin interface characterization and modeling for surface biopotential recording: Impedance measurements and noise," In 2013 2nd International Conference on Advances in Biomedical Engineering, pp.49–52. IEEE.
9. T. Kato, A. Ueno, S. Kataoka, H. Hoshino and Y. Ishyama, "An application of capacitive electrode for detecting electrocardiogram of neonates and infants," n 2006 International Conference of the IEEE Engineering in Medicine and Biology Society, pp.916–919. IEE.ISO 690.
10. D. P. Subha, P. K. Joseph, R. Acharya, and C. M. Lim, "Eeg signal analysis: a survey," Journal of medical systems, vol. 34, 2010, pp.195–212.
11. S. Sinha, L. Sullivan, D. Sabau, D. S. J, E. Dombrowski, J. Halford, A. Hani, F. Drislane, and M. Stecker, "American clinical neurophysiology society guideline 1: minimum technical requirements for performing clinical electroencephalography," The Neurodiagnostic Journal, vol. 56, 2016, pp.235–244.

12. T. Ragheb, and L. Geddes, “Electrical properties of metallic electrodes, “Electrical properties of metallic electrodes,” Medical and Biological Engineering and computing, vol. 28, 1990, pp.182.
13. A. Geddes, “ Electrodes and the Measurement of Bioelectrical Events,” WILEY-INTERSCIENCE, 1972, pp.99–102.
14. L. Peng, J. Liu, Y. Dong, B. Yang, X. Chen, and C. Yang,“Parylene-based flexible dry electrode for bioptential recording,” Sens. Actuators B Chem, vol. 231, 2016, pp.1–11.
15. S. Rodrigues, P. Fiedler, N. Küchler, R. Domingues, C. Lopes, J. Borges, J. Haueisen, and F. Vaz, “,Dry Electrodes for Surface Electromyography Based on Architectured Titanium Thin Films. Materials,” Materials, vol.13, 2020, pp.2135.
16. H. Hinrichs, M. Scholz, A. Baum, J. Kam, R. Knight, and H. Heinze,“Comparison between a wireless dry electrode EEG system with a conventional wired wet electrode EEG system for clinical applications. Scientific Reports,” Scientific Reports, vol.10, 2020, pp.1–14.
17. J. Liu, X. Liu, E. He, F. Gao, Z. Li, G. Xia, S. Xu, and X. Cai, “novel dry-contact electrode for measuring electroencephalography signals,” Sensors and Actuators A: Physical, vol. 294, 2019, pp.73–80.
18. M. Mañanas, S. Romero, Z. Topor, E. Bruce, P. Houtz, and P. Caminal, “ Cardiac interference in myographic signals from different respiratory muscles and levels of activity,” In 2001 Conference Proceedings of the 23rd Annual International Conference of the IEEE Engineering in Medicine and Biology Society, vol. 2, 2001, pp. 1115–1118.
19. C. Pylatiuk, M. Muller, A. Kargov, S. Schulz. O. Schill, M. Reischl, and, G. Brettahuer, “Comparison of surface EMG monitoring electrodes for long-term use in rehabilitatios device control,” In 2009 IEEE International Conference on Rehabilitation Robotics, 2009, pp.300–304.
20. H.Jung, J. Moon, D. Baek, J. Lee, Y. Choi, J. Hong, S. Lee, “CNT/PDMS composite flexible dry electrodesfor long-term ECG monitoring,” IEEE Trans. Biomed. Eng, vol. 59, 2012, pp.1472–1479.
21. P Alves, S. Pinto, H. de Sousa, and M. Gil, “Surface modification of a thermoplastic polyurethane by low-pressure plasma treatment to improve hydrophilicity,” Journal of Applied Polymer Science, vol. 122, 2011, pp.2302–2308.
22. R.Tyler, J. Choi, J. Bandodkar, S. Krishnan, P. Gutruf, L. Tian, R. Ghaffari, and J. Rogers, “ Bio-integrated wearable systems: a comprehensive review,” Chemical reviews, vol.119, 2019, pp.5461–5533.
23. H. Peng, J. Dong, B. Yang and X. Chen, “Parylene-based flexible dry electrode for bioptential recording,” Sensors and Actuators B: Chemical,vol.231, 2016, pp.1-11.
24. L. Wang, J. Liu, X. Yan, B. Yang, and C. Yang, “MEMS-based pyramid microneedle electrode for long-term EEG measurement,” Microsystem technologies, vol.19, 2013, pp.269–276.
25. R. Wang, X. Jiang, W. Wang, and Z. Li, “A microneedle electrode array on flexible substrate for long-term EEG,” Sensors and Actuators B: Chemical, vol. 244, 2017, pp.750-758.
26. L. Ren, S. Xu, J. Gao, Z. Lin, Z. Chen, B. Liu, L. Liang, and L. Jiang, “Fabrication of Flexible Microneedle Array Electrodes for Wearable Bio-Signal Recording,” Sensors, vol.18, 2018, pp.1191.
27. Y. Chen, M. Beeck, L. Vanderheyden, E. Carrette, V. Mihajlović, K. Vanstreels, B. Grundlehner, S. Gadeyne, P. Boon, and C. V. Hoof, “Soft, Comfortable Polymer Dry Electrodes for High Quality,“, vol.14, 2014, pp.23758–23780.

28. K. Gao, H. Yang, X. Wang, B. Yang, and J. Liu,“Soft pin-shaped dry electrode with bristles for EEG signal measurements,” Sensors and Actuators A: Physical, Vol.283, 2018, pp.348–361.
29. S. Rodrigues, P. Fiedler, N. Küchler, R. Domingues, C. Lopes, J. Borges, J. Haueisen, and F. Vaz, “,Dry Electrodes for Surface Electromyography Based on Architectured Titanium Thin Films. Materials,” Materials, vol.13, 2020, pp.2135.
30. S. Ghosh, M. Mahadevappa, and J. Mukhopadhyay,“A 2D electrode-skin model for electrical and contact impedance characterization of Bio Impedance,” In 2016 IEEE Region 10 Conference (TENCON) pp.2292–2295. IEEE.
31. G. Anand, A. Lowe, and A. Al-Jumaily, “Simulation of impedance measurements at human forearm within 1 kHz to 2 MHz,” Journal of Electrical Bioimpedance, vol.7, 2019, pp.20–27.
32. F. Lu, C. Wang, R. Zhao, L. Du, Z. Fang, X. Guo, and Z. Zhao, “Review of stratum corneum impedance measurement in non-invasive penetration application,” Biosensors, vol 8, 2018, pp.31.
33. U. Birgersson, E. Birgersson, P.Åberg, I. Nicander, and S. Ollmar, “,Non-invasive bioimpedance of intact skin: mathematical modeling and experiments,” Physiological measurement, 2010, vol.32, p.1.
34. A. Searle, and L. Kirkup, “A direct comparison of wet, dry and insulating bioelectric recording electrodes,” Physiological measurement, vol.21, 2000, pp.271.
35. P. Griss, P. Enocksson, H. Tolvanen, P. Merlainen, S. Ollmar, and G. Stemme, “Spiked biopotential electrodes,” n Proceedings Ieee Thirteenth Annual International Conference On Micro Electro Mechanical Systems, pp.323–328. IEEE
36. J. Baek, H. An, M. Choi, S. Park, and H. Lee, “Flexible polymeric dry electrodes for the long-term monitoring of ECG,” Sensors and Actuators A: Physical, vol.143, 2008, pp.423–429.
37. T. Ragheb, and L. Geddes, “Electrical properties of metallic electrodes, “Electrical properties of metallic electrodes,” Medical and Biological Engineering and computing, vol. 28, 1990, pp.182.
38. J. Kam, S. Griffin, S. Patel, H. Hinrichs, J. Heinze, adn T. Knington, “Systematic comparison between a wireless EEG system with dry electrodes and a wired EEG system with wet electrodes,” NeuroImage, vol.184, 2019, pp.119–129.
39. P. Fiedler, J. Haueisen, D. Jannek, S. Griebel, L. Zentner, F. Vaz, and C. Fonseca, “Comparison of three types of dry electrodes for electroencephalography,” Acta Imeko, vol.3, 2014, pp.33–37.
40. A. Yeung, H. Garudadri, C. Van Toen, P. Balkan, S. Makeig, and N. Virji-Babul, “Comparison of foam-based and spring-loaded dry EEG electrodes with wet electrodes in resting and moving conditions,” In 2015 37th Annual International Conference of the IEEE Engineering in Medicine and Biology Society, pp.7131–7134
41. T. McAdams, J. Jossinet, A. Lackermeier, and F. Risacher, “Factors affecting electrode-gel-skin interface impedance in electrical impedance tomography,” Medical and Biological Engineering and Computing, vol.34,1996, pp.397–408.
42. E.Huigen, A. Peper, and A. Grimbergen, “Investigation into the origin of the noise of surface electrodes. Medical and biological engineering and computing,” Medical and biological engineering and computing, vol 40. 2002, pp.332–338.