Sensors and Actuators A 174 (2012) 96–102

Contents lists available at SciVerse ScienceDirect

Sensors and Actuators A: Physical

journal homepage: www.elsevier.com/locate/sna

A 3D printed dry electrode for ECG/EEG recording

P. Salvo[a,*], R. Raedt[b], E. Carrette[b], D. Schaubroeck[a], J. Vanfleteren[a], L. Cardon[c,d]

[a] Centre for Microsystems Technology (CMST), Faculty of Engineering, University of Ghent, 914A Technologiepark, B-9052 Ghent-Zwijnaarde, Belgium
[b] Department of Neurology, Ghent University Hospital, 185 De Pintelaan, 9000 Ghent, Belgium
[c] Faculty of Applied Engineering Sciences, CPMT Research Group, University College Ghent, Voskenslaan 362, B-9000 Ghent, Belgium
[d] Department of Materials Science and Engineering, University Ghent, Technologiepark 903, B-9052 Ghent, Belgium

ARTICLE INFO

Article history:
Received 4 May 2011
Received in revised form 8 December 2011
Accepted 8 December 2011
Available online 17 December 2011

Keywords:
3D printed dry electrode
Bioelectric signal recording
Electrocardiogram (ECG)
Electroencephalogram (EEG)

ABSTRACT

In this paper, the design, fabrication and testing of a 3D printed dry electrode is proposed. 3D printing represents an authentic breakthrough for the development and mass production of dry medical electrodes. In fact, it allows a fast and low cost production of high precision tridimensional shapes. This technique is reliable and efficient, and facilitates controllability over the whole process. Initially, 3D capable design software is used to draw the electrode model. The resulting file is simply loaded in a 3D printer whose resolution is 42 μm on *x*- and *y*-axes, and 16 μm on *z*-axis. The electrode is made by an insulating acrylic-based photopolymer. It consists of 180 conical needles (distance = 250 μm) on a truncated conical base. The metallization process undergoes two steps: sputtering of titanium as adhesion promotion layer and evaporation of gold to lower the impedance and prevent oxidation of the electrode. After electrode characterization, experimental results are presented and compared with planar wet Ag/AgCl electrodes for recording ECG–EEG.

© 2011 Elsevier B.V. All rights reserved.

1. Introduction

Standard planar Ag/AgCl electrodes suffer from two major drawbacks: skin preparation, which produces abrasive lesions in some subjects, and the application of conductive gel or paste, which is necessary to reduce electrode-to-skin impedance mismatch. The patient can be sensitive to gel, thus allergic reaction, rash or skin irritation can occur. For long term monitoring, due to change in humidity, the variation in the gel impedance causes artifacts. Furthermore, skin preparation and gel application are time consuming when a high number of electrodes is required, e.g. for epilepsy monitoring. The demand for more comfortable and user-friendly electrodes has led to the development of an increasing number of dry devices that can overcome the limitations of wet electrodes. Dry electrodes have the disadvantages of higher electrode-skin impedance, additional circuitry and susceptibility to movement artifacts. However, in literature it is reported that, because of the accumulation of perspiration under the electrodes, after a settling time, the impedance greatly decreases, and artifacts noise is lower than for wet electrodes [1]. Several materials and methods for fabrication of dry electrodes have been explored for the application [2–8]. Dry electrodes can be made by semiconductor, metal or metal-filled polymers and soft, flexible or hard types are reported. Examples of production of sub-dermal 3D spiked arrays by micromachinery have been published [3,7]. In this paper we present 3D printing for rapid-prototyping, a new technique that can be applied for surface electrodes mass production with high yield. In fact, within the limits of the printer resolution and software capabilities, it allows fabrication of a complex 3D electrode in a short time and with few efforts. Electrodes with complex 3D geometries could facilitate recording electrophysiological activities in presence of hair, which has to be considered for a correct measurement [2]. The development of hair-compatible electrodes would greatly improve the patient's comfort. It would be possible to avoid shaving the chest in order to record an ECG if a large amount of hair is present [9]. Similarly, in electroencephalography, the long and laborious procedure to remove gel from hair would not be necessary anymore, and, if integrated in a proper cap, electrodes could be secured to patients' head without any glue.

Properties of 3D printed electrodes are discussed in detail and their ability to record electrocardiographic and electroencephalographic signals are compared to planar Ag/AgCl electrodes.

2. Materials and methods

2.1. Design

The 3D electrode model is designed in AutoCAD® environment. The electrode consists of a truncated conical support (0.5 mm high, bottom base diameter = 15 mm, top base diameter = 13 mm) on

* Corresponding author. Tel.: +32 9 264 53 54; fax: +32 9 264 53 74.
E-mail address: pietro.salvo@gmail.com (P. Salvo).

0924-4247/$ – see front matter © 2011 Elsevier B.V. All rights reserved.
doi:10.1016/j.sna.2011.12.017

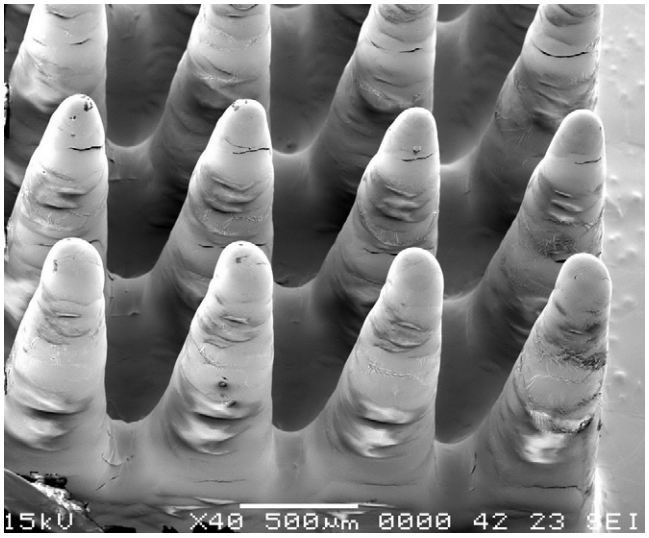

Fig. 1. Partial SEM image of electrode tilted by 35°.

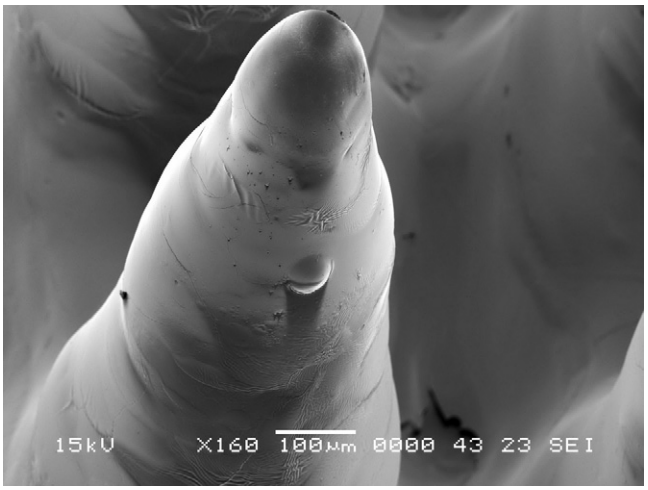

Fig. 2. SEM image of a 3D printed needle tilted by 45°.

which an array of conical needles is placed. Each needle is 3 mm high with base diameter of 600 μm and a tip diameter of about 100 μm. The diameter of the hair fiber varies from approximately 50 μm to 100 μm [10]. Thus, 100 μm has to be assumed as the lower limit for the needle distance. The needles could pierce the skin if the contact resistance is too high. A possible solution is increasing the contact area. In fact, like in a bed of nails, if the pressure is distributed evenly over a high density array of needles, the patient's skin will not be penetrated. After several tests, a good compromise for the needle distance (i.e. the distance between two bases) is found to be 250 μm. This choice leads to an array made up by 180 needles. In case of very long hair, needles can be 4 mm and 5 mm high. It is not recommended to use higher needles as, if their base length is kept constant to preserve their number (i.e. 180), they imply sharper tips. Therefore, there would be a concrete risk to damage the skin.

The AutoCAD® project is eventually exported from dwg to stl format, which is the standard file type for the fabrication process described in next sub-section.

2.2. 3D printing

3D printing for rapid prototyping is a manufacturing method where consecutive horizontal thin layers of polymeric material are jetted to create a tridimensional object. We use the Eden^{TM} 350V printer by Objet Geometries. Resolution is 16 μm along z-axis and 42 μm along x- and y-axes. Net build size is 340 mm × 340 mm × 200 mm; therefore, it is suitable for electrode mass production as approximately four hundred and fifty electrodes, 0.1 mm distant from each other, can be fabricated in a single run. Objet PolyJet^{TM} technology enables to jet 16 μm thick horizontal layers of a photopolymer. The chosen material, FullCure® 720, is an insulating transparent biocompatible acrylic-based resin. Datasheet reports tensile strength of 60.3 MPa, elongation at break of 20%, flexural modulus of 1718 MPa and shore hardness of 83 (scale D). We measured the glass transition temperature (T_g) by differential scanning calorimetry (DSC) that returns a value of roughly 50 °C. After jetting, each layer is cured by ultraviolet (UV) light. Following the printing process, the gel-like support material is removed by dipping the model in 5–10% NaOH solution for 10 min; afterwards, it is dried at 27 °C for 10 min. The model can be handled and used immediately, without post-curing. Figs. 1 and 2 show scanning electron microscope (SEM) images of the model.

2.3. Metallization

FullCure® 720 has been tested by thermo-gravimetric analysis (TGA) to determine weight loss in relation to temperature change. TGA is performed under ambient conditions with a heating rate of 10 °C/min. Results are shown in Fig. 3.

With respect to the initial value at room temperature, the weight is 99.9% at 83 °C, 99.5% at 143 °C and 99% at 177 °C. Therefore, in order to maintain the material integrity, processing steps can be considered acceptable at temperatures below 150 °C. Metallization process is carried out in two consecutive steps in a vacuum coating chamber (Univex 450 by Leybold). To avoid presence of dust particles and machine contamination, electrodes are cleaned with isopropyl alcohol and dried at 27 °C for 10 min. During first step, an adhesion promotion layer of titanium (150 nm) is sputtered. In the same vacuum cycle, a gold layer (250 nm) is evaporated to lower the impedance and to prevent both corrosion and oxidation. Eventually, a gold plated connector for lead wire, Multi-Contact SLS205-BF (DIN 42802, diameter = 2 mm), is glued to the bottom by a conductive silver paste (Emerson and Cuming, CE 3104 WXL) and cured at 120 °C for 10 min. An illustrative example of a 3D printed electrode is shown in Fig. 4.

Gold adhesion is confirmed by scotch tape test with excellent results. In fact, the coating is not pulled off and no appreciable differences in the tape are noticed in the image taken by an optical microscope (Nikon Optiphot 200) after removing it.

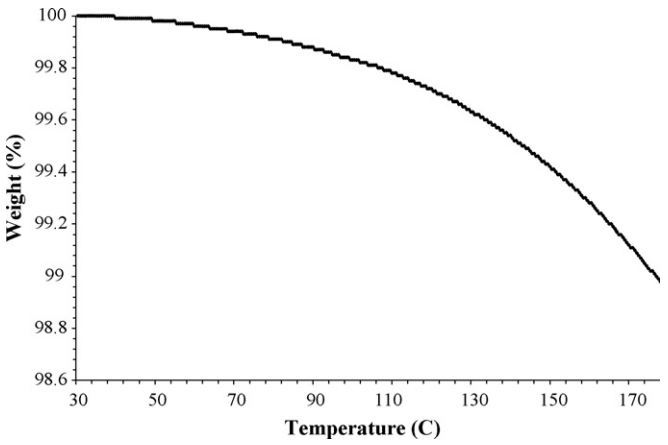

Fig. 3. Thermo-gravimetric analysis of Fullcure® 720. The weight loss is 0.1% at 83 °C, 0.5% at 143 °C and 1% at 177 °C.

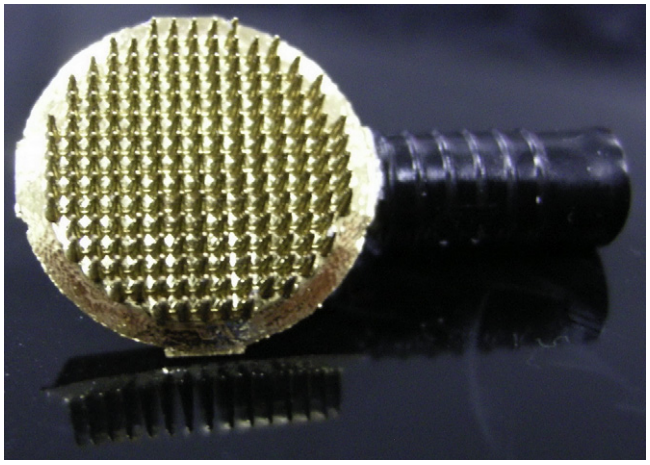

Fig. 4. 3D printed electrode in its final form.

The method described in [11] and [12] is used to measure electrode-skin impedance (ESI). As signals of interests are ECG and EEG, ESIs are measured at chest and forehead as indicated in Fig. 5.

For the chest, A and C are wet electrodes Kendall™ ARBO (type H135SG, diameter 35 mm). B is the dry electrode under test. By means of Agilent 33220A, a sinusoidal voltage, V_S, is applied between A and B. The current flowing through resistor R (100 KΩ), i_R, is

$$i_R = \frac{V_{aB}}{R}$$

where V_{aB} is the voltage drop across R. Ideally, no current i_{BC} flows through between B and C, therefore voltage V_{BC} is only due to the voltage drop across Z_B, which is the ESI to be measured, and the following relation holds:

$$Z_B = \frac{V_{BC}}{i_R}$$

The voltage generator is set to supply a maximum current of 2 μA peak to peak. Peak to peak voltages V_{aB} and V_{BC} are buffered by an instrumentation amplifier (Linear Technologies LT1167), and measured by an oscilloscope (Tektronics TDS55104B). The high LT1167 input impedance (1 TΩ typical and 200 GΩ minimum) allows assuming i_{BC} equal to zero. For the forehead, $Z_{B'}$ is evaluated with the same procedure. The magnitude of impedance for five repeated measurements, each starting 5 min after application of the electrodes, is illustrated in Figs. 6 and 7. The averages of standard deviations in the frequency span are 100 Ω and 75 Ω at chest and forehead, respectively.

Such large values are expected for dry electrodes [13]. To convert from high to low impedance signal, a pre-amplification stage (usually a buffer amplifier) can be inserted before the measurement unit [14]. This solution is referred to as active electrode.

3. Results

3.1. ECG

3-Lead ECG is recorded by a lab-made device ($f_{sampling}$ = 170 Hz) and digitally filtered with low and high pass filters ($f_{LPcut\text{-}off}$ = 0.5 Hz and $f_{HPcut\text{-}off}$ = 40 Hz, respectively). An instrumentation amplifier (Analog Devices AD627, input impedance = 20 GΩ) is employed as input stage for the electrocardiogram. Wet electrodes are Kendall™ ARBO (type H135SG, diameter 35 mm). Fig. 8 shows the ECGs for a subject with a moderate hairy chest.

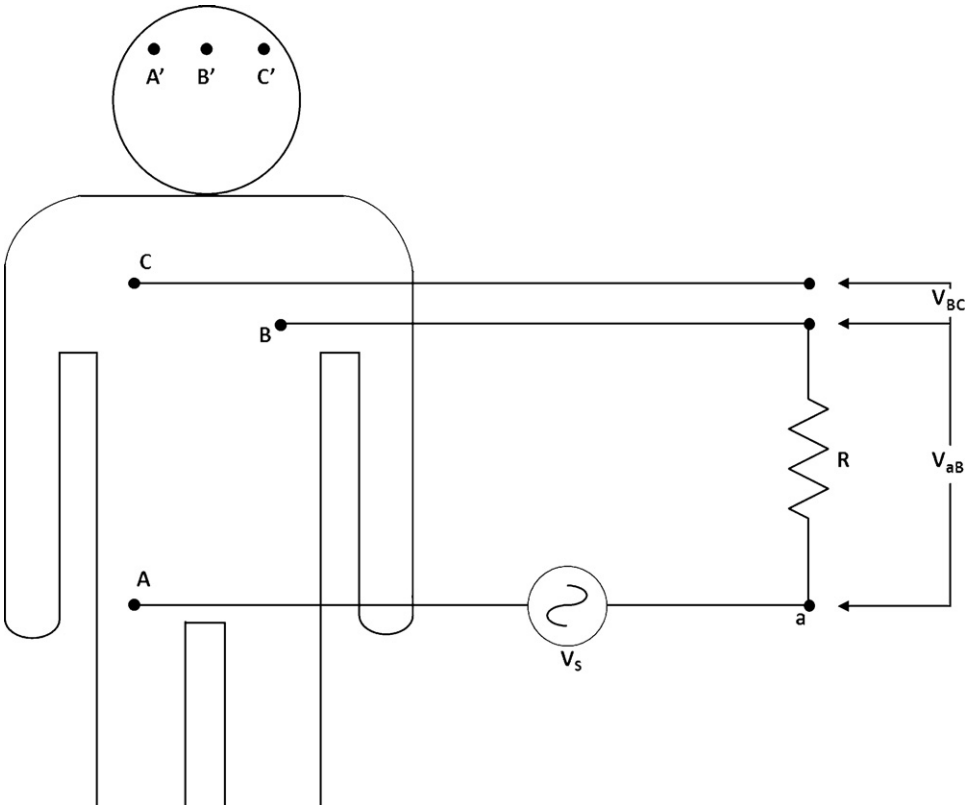

Fig. 5. Electrode-skin impedance measurement at chest and forehead. Dry electrodes are placed in B and B′. A, A′, C and C′ are wet electrodes. The circuit employed to measure Z_B is used for Z'_B as well.

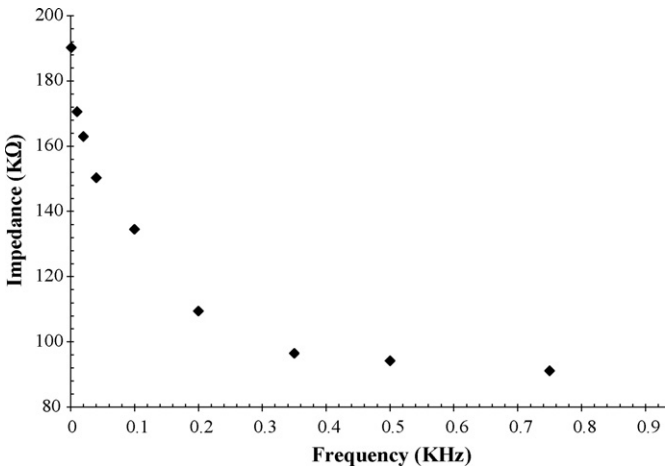

Fig. 6. Electrode-skin impedance for the chest.

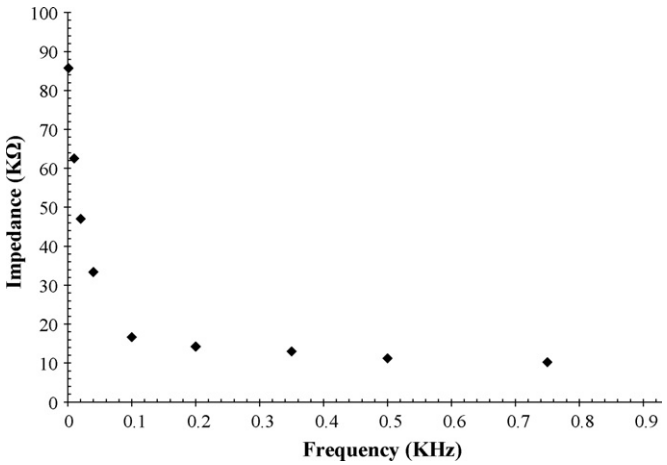

Fig. 7. Electrode-skin impedance for the forehead.

The two waveforms look very similar. Both ECGs present low frequency noise due to motion artifacts that can be suppressed with appropriate filtering techniques and recording devices. Another possibility is to analyze the signals in the frequency domain and isolate the band of interest. Thus, to corroborate the visual inspection, power spectral densities (PSDs) are calculated by Welch's method implemented in Matlab, using $f_{sampling}$ to compute the frequency vector (Fig. 9).

Murthy et al. [15] report that in a typical QRS complex of normal duration, peak power is localized in the range from 4 to 12 Hz. In such interval, the difference in average power, $\Delta_{dry-wet}$, evaluated via a rectangle approximation of the integral of PSDs, is extremely small (3.2×10^{-4} dB). Therefore, the power spectral analysis confirms the reliability of 3D printed dry electrodes for ECG recording.

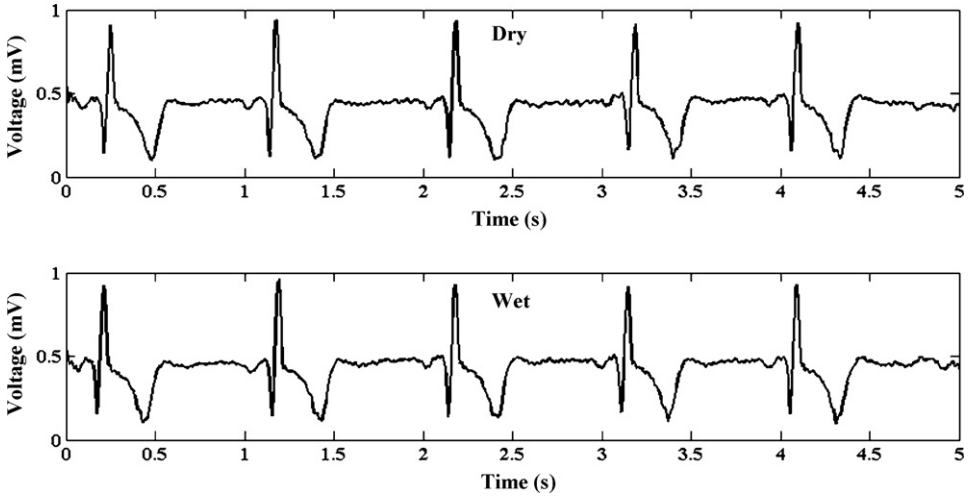

Fig. 8. ECGs from (a) dry and (b) wet electrodes.

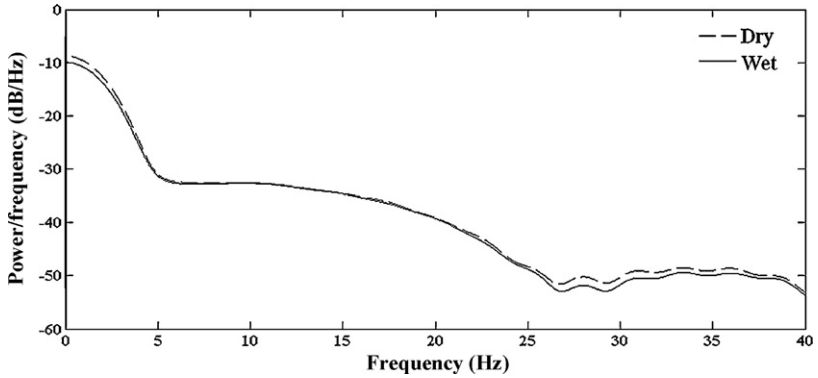

Fig. 9. Welch power spectral density estimate for dry electrodes (dashed line) and wet electrodes (solid line).

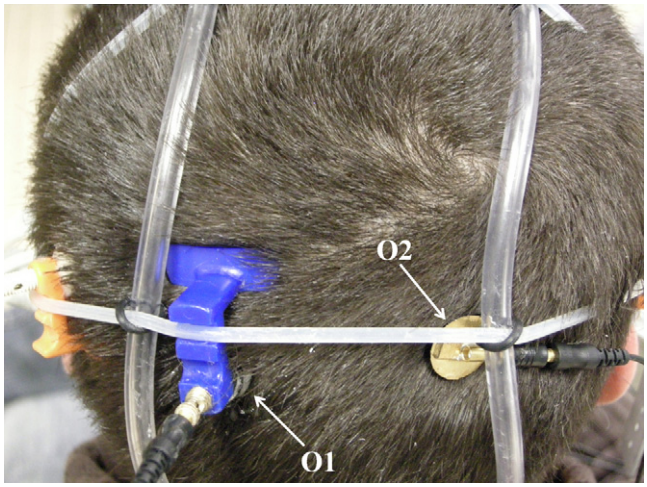

Fig. 10. Dry and bridge electrodes placed in position O2 and O1, respectively.

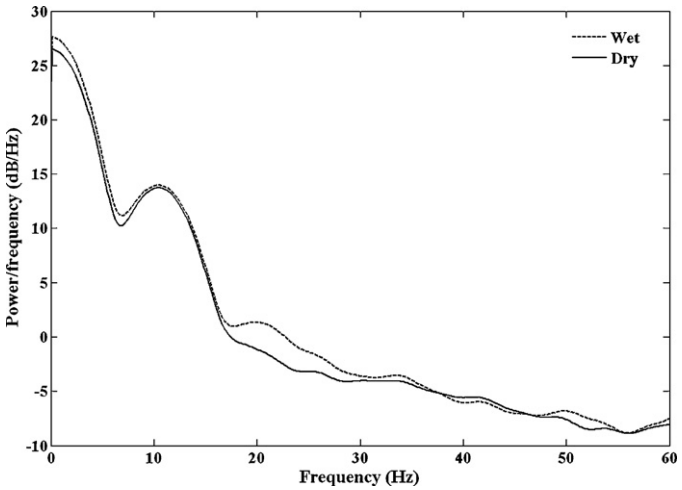

Fig. 12. Welch power spectral density estimates for bridge electrodes (dashed line – position O1) and dry (solid line – position O2).

3.2. EEG

EEG tests are carried out at Ghent University Hospital and we used the same electrodes chosen for the ECG recording in order to prove them reusable. Electrodes are connected to a Micromed Brain Quick unit with a 32 channels smart acquisition module (CMRR > 110 dB, noise < 0.16 μV_{rms}) and held in place on the subject's head by means of an elastic net. System settings are $f_{sampling}$ = 256 Hz, $f_{HPcut\text{-}off}$ = 0.1 Hz and $f_{LPcut\text{-}off}$ = 67 Hz. To lower the impedance of dry electrodes, a buffer stage, not integrated in the electrode, has been realized (main component is Texas Instruments OPA2336).

3D printed electrodes are placed at positions O2 (Fig. 10) and Fp2 and compared to bridge Belek Ag/AgCl electrodes placed at positions O1 and Fp1 during wakeful relaxation with closed eyes. This condition promotes the occurrence of alpha rhythms (8–13 Hz waves), which are higher in amplitude on the dominant side of the head [16]. As reference, a dry electrode is placed at position T6. The Micromed Unit has a built-in function to determine the impedance for each electrode. Acceptable values are below 10 kΩ. Brain Quick returns 1–3 kΩ and 2–5 kΩ at Fp1 and Fp2, respectively, whereas 4–7 kΩ and 1–3 kΩ are found at O1 and O2, respectively. These values testify the good performance in hair penetration of dry electrodes. Fig. 11 shows the alpha waves at O2 compared to its symmetric O1. PSDs are shown in Fig. 12.

In Fig. 12, the typical alpha wave peak is clearly visible. The average powers calculated in this band are 101 dB and 108 dB for dry and bridge electrodes, respectively. Figs. 13 and 14 show the acquired signal at Fp1 and Fp2.

In the 8–13 Hz range, the average powers are 50 dB and 49 dB at Fp1 and Fp2, respectively. The same values have been obtained for electrodes with 4 mm and 5 mm needles. These results support comparable functioning of dry and bridge Ag/AgCl electrodes.

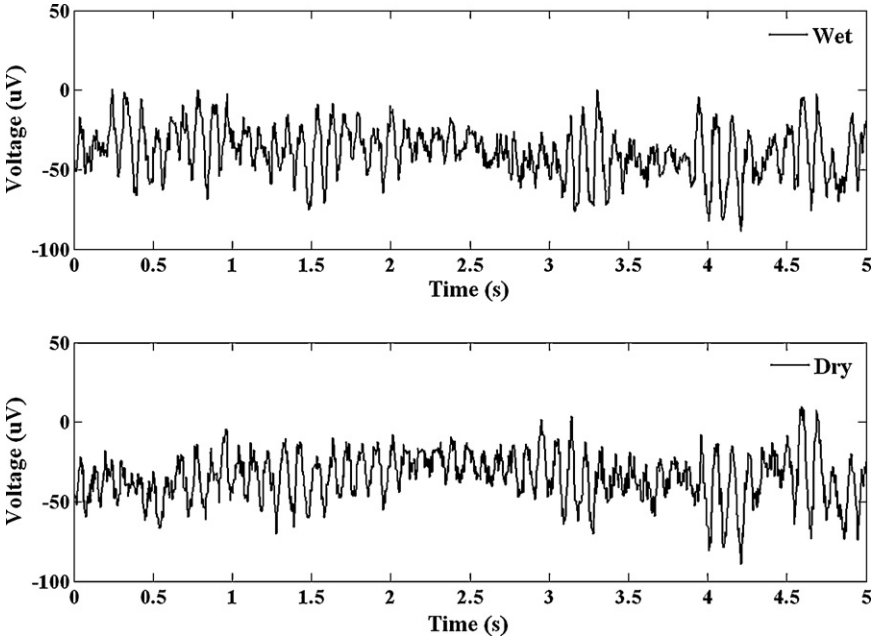

Fig. 11. Alpha waves for bridge electrodes (position O1) and dry (position O2).

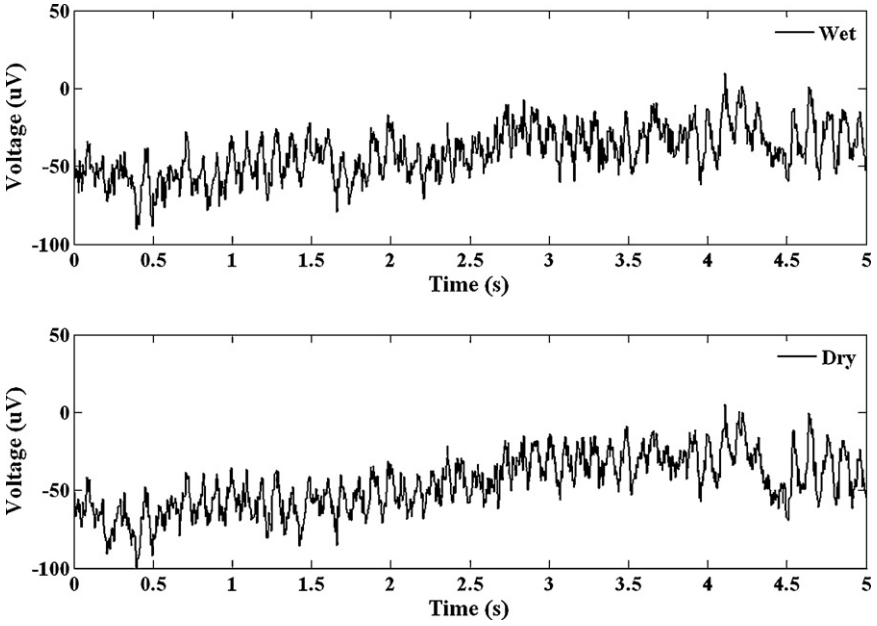

Fig. 13. EEG recorded by bridge (position Fp1) and dry electrodes (position Fp2).

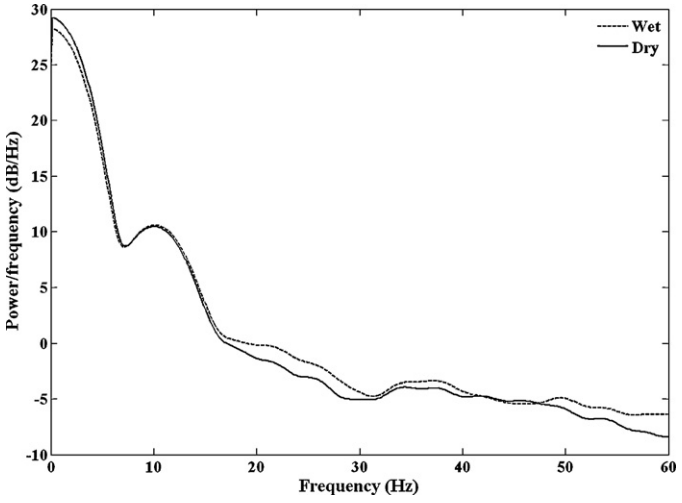

Fig. 14. PSDs for bridge electrodes (dashed line – position Fp1) and dry (solid line – position Fp2).

4. Conclusion

An ongoing challenge in the domain of medical electrodes is the development of dry sensors able to substitute the standard Ag/AgCl type to improve patient's comfort. In this paper, the prototype of a 3D printed dry electrode for ECG/EEG recording is proposed. No skin preparation and no electrolytic gel are needed prior to the application of the dry electrodes on the skin. The production differs from previous works in array fabrication as 3D printing does not require micro-machinery or silicon processing. This approach enables a rapid and low cost fabrication of the desired electrode geometry, with high resolution features in a reproducible way, and it can be employed for mass production. Furthermore, to our knowledge, for the first time a reliable method to metallize the model surface is described. In fact, titanium and gold coatings have proved to adhere successfully towards the acrylic resin Fullcure® 720. This sensor has suitable physical and mechanical properties for daily manipulation and medical use. The electrode mechanical strength does not require specific care to handle it. Remarkable advantages are that it does not require any sub-dermal insertion, it can be easily cleaned, it is reusable and it can be plugged to standard wires.

Preliminary comparative tests show that 3D printed electrodes can be applied without gel and skin preparation. They provide results comparable to wet electrodes in electrocardiography when coupled with a high input impedance amplifier integrated in the acquisition board. Power spectral densities show negligible variations in the QRS complex frequency interval. Analysis of power spectra densities in EEG tests has led to results analogous to Ag/AgCl electrodes. However, even though these conclusions are promising, further tests are necessary to validate the efficacy of this technology. In fact, it is known that current can cross the electrolyte–electrode interface, created by skin perspiration, as cations and electrons are formed on electrode oxidation. The cations are discharged into the electrolyte whereas the electrons flow through the lead wires. The consequence is the generation of a voltage known as the half-cell potential, which causes a DC offset in the EEG. The magnitude depends on the type of metal and can be highly variable [17]. Thus, when EEG leads attached to a patient occur to be made of different materials, there could be a substantial voltage between them that can affect the comparison. Future improvements are the integration of the buffer stage in the electrode structure for better portability and the reduction of external interferences in the band of clinical interest.

For EEG recording, different caps need to be tested. In principle, artifacts due to inadequate contact with skin could be removed by tightening the cap. Nevertheless, this method could be unpleasant and even painful for the patient. Instead, it is believed that an *ad hoc* cap with integrated electrodes will ensure a good contact with the skin. If the previous recommendations are followed, the proposed dry electrodes can reveal to be suitable as valid sensors for ECG and EEG recording.

Acknowledgments

This work was supported by the Research Fund of the University of Ghent (BOF-GOA). The authors would like to thank Prof. A. Van Calster from CMST, University of Ghent, and the RPT research group from University College Ghent for their contributions to this work.

References

[1] A. Searle, L. Kirkup, A direct comparison of wet, dry and insulating bioelectric recording electrodes, Physiol. Meas. 21 (2000) 271–283.
[2] A. Karilainen, J. Hansen, S. Muller, Dry and capacitive electrodes for long-term ECG-monitoring, in: Proceedings of the 8th Annual Workshop on Semiconductor Advances for Future Electronics, Veldhoven, 2005, pp. 155–161.
[3] P. Griss, H.K. Tolvanen-Laasko, P. Meriläinen, G. Stemme, Characterization of micromachined spiked biopotential electrodes, IEEE Trans. Biomed. Eng. 49 (6) (2002) 597–604.
[4] K.P. Hoffmann, R. Ruff, Flexible dry surface-electrodes for ECG long-term monitoring, in: Engineering in Medicine and Biology Society, Proceedings of the 29th Annual International Conference of the IEEE, EMBS, Lyon, 2007, pp. 5739–5742.
[5] C. Fonseca, J.P. Silva Cunha, R.E. Martins, V.M. Ferreira, J.P. Marques de Sá, M.A. Barbosa, A. Martins da Silva, A novel dry active electrode for EEG recording, IEEE Trans. Biomed. Eng. 54 (1) (2007) 162–165.
[6] J.Y. Baek, J.H. An, J.M. Chio, K.S. Park, S.H. Lee, Flexible polymeric dry electrodes for the long-term monitoring of ECG, Sens. Actuators A 143 (2008) 423–429.
[7] W.C. Ng, H.L. Seet, K.S. Lee, N. Ning, W.X. Tai, M. Sutedja, J.Y.H Fuh, X.P. Li, Microspike EEG electrode and the vacuum-casting technology for mass production, J. Mater. Process. Technol. 209 (2009) 4434–4438.
[8] M.R. Neuman, Biopotential Electrodes, in: J.D. Bronzino (Ed.), The Biomedical Engineering Handbook, vol. 2, ch. 48, 2nd ed., CRC Press LLC, Boca Raton, 2000.
[9] T. Phalen, B. Aehlert, 12-Lead ECG in Acute Coronary Syndromes, Elsevier Health Science, 2005, 42 pp.
[10] B. Bhushan, G. Wei, P. Haddad, Friction and wear studies of human hair and skin, Wear 259 (2005) 1012–1021.
[11] G.E. Bergey, R.D. Squires, W.C. Sipple, Electrocardiogram recording with pasteless electrodes, IEEE Trans. Biomed. Eng. 18 (3) (1971) 206–211.
[12] I. Zepeda-Carapia, A. Marquez-Espinoza, C. Alvarado-Serrano, Measurement of skin electrode Impedance for a 12-lead electrocardiogram, in: Proceedings of the 2nd International Conference on Electrical and Electronics Engineering (ICEEE) and XI Conference on Electrical Engineering (CIE 2005), Mexico City, Mexico, September 7–9, 2005.
[13] A. Baba, M.J. Burke, Electrical characterisation of dry electrodes for ECG recording, in: Proceedings of the 12th WSEAS International Conference on Circuits, Heraklion, Greece, July 22–24, 2008.
[14] T.C. Ferree, P. Luu, G.S. Russell, D.M. Tucker, Scalp electrode impedance, infection risk, and EEG data quality, Clin. Neurophysiol. 112 (2001) 536–544.
[15] V.K. Murthy, T.M. Grove, G.A. Harvey, L.J. Haywood, Clinical usefulness of ECG frequency spectrum analysis, in: Proceedings of the Computer Application in Medical Care, Washington, DC, 1978, pp. 610–612.
[16] D.P. Subha, P.K. Joseph, U.R. Acharya, C.M. Lim, EEG signal analysis: a survey, J. Med. Syst. 34 (2010) 195–212.
[17] P. Tallgren, S. Vanhataloa, K. Kailaa, J. Voipio, Evaluation of commercially available electrodes and gels for recording of slow EEG potentials, Clin. Neurophysiol. 116 (2005) 799–806.

Biographies

Pietro Salvo received the M.Sc. degree in Electronics Engineering and the Ph.D. degree in Automation, Robotics and Bioengineering from University of Pisa, Italy, in 2004 and 2009, respectively. He is currently a post-doctoral researcher at Centre for Microsystems Technology (CMST), University of Ghent, Belgium, where he joined in 2009. He was a research associate at Institute for the Conservation and Promotion of Cultural Heritage, at National Council of Research, Florence, Italy, from 2008 until 2009. His actual research interests mainly include the study of sensors and microfluidic systems for bioengineering and biomedical applications, the design and fabrication of 3D dry electrodes for electrocardiography and electroencephalography recording and monitoring, and the development of an inertial measurement unit for tracking fast human movements.

Robrecht Raedt is an Assistant Research Professor in Intracranial Electrophysiology, working at the Laboratory for Clinical and Experimental Neurophysiology of Ghent University in Belgium. In 2002 he obtained a Masters degree in Biology at Ghent University. In 2007 he successfully completed his Ph.D. research at the Faculty of Medicine and Health Sciences of Ghent University with his research on cell therapy for epilepsy. Currently, his research focuses on the optimization and mechanism of action of neurostimulation (vagus nerve stimulation and deep brain stimulation) for epilepsy both in animal models as in human with epilepsy.

Evelien Carrette received the M.Sc. degree in Biomedical Sciences and the Ph.D. degree in Medical Sciences from Ghent University, Belgium, in 2006 and 2011 respectively. She is currently a post doctoral researcher at the Laboratory for Clinical and Experimental Neurophysiology (LCEN) and the Reference Center for Refractory Epilepsy (RCRE), Ghent University (Hospital), Belgium, where she stayed following the termination of her Ph.D. thesis. In this position she works as the Research Coordinator within the Ghent Institute for Neuroscience and as Policy Officer at the Neurology Department and Center for Neurophysiological Monitoring at Ghent University Hospital since October 2011. Her research interests include refractory epilepsy, the presurgical evaluation of patients with refractory epilepsy, neurostimulation and neuromodulation, electroencephalography (EEG) both non-invasively and invasively recorded and magnetoencephalography (MEG).

David Schaubroeck was born in 1980 in Belgium. He obtained a M.Sc. degree in Chemistry at the University of Ghent (Belgium) in 2004. Until 2006, he worked as a scientific researcher for a spin-off company of Ghent University specialized in the development of olefin metathesis catalysts. Thereafter, he joined the research group of Prof. P. Van Der Voort at Ghent University where he developed ordered mesoporous polymers. Currently, he is a Ph.D. student at the electronics and information department (Ghent University, Belgium) under the supervision of Prof. André Van Calster and Prof. Peter Dubruel. His research interests include polymer materials for electronics, biomaterials, adhesion and polymer surface modification.

Jan Vanfleteren obtained his Ph.D. in electronic engineering from Ghent University (Belgium) in 1987. He is currently a senior engineer at the IMEC-CMST group and is involved in the development of novel interconnection, assembly and substrate technologies, especially in wearable electronics technologies. As a project manager for CMST he has a long standing experience in co-ordination and co-operation in EC funded projects. In 2004 he was appointed as part time professor at the Ghent University. He is a member of IMAPS and IEEE and (co)-author of over 200 papers in international journals and conferences and he holds 14 patents/patent applications.

Ludwig Cardon became Mechanical Engineer in 1988 and joined the University College of Ghent, Belgium since 1990. He received his Ph.D. degree in Engineering in 2006 from the Birmingham City University, UK. Since 2006, he is the head of the Centre for Polymer & Material Technologies CPMT and professor at the faculty of Applied Engineering Sciences of the University College of Ghent. He is co-founder of the rapid prototyping research group (RPT), now integrated in CPMT. His actual research interests mainly include high-tech polymer technology/processing and RPT for several applications such as technical parts, biobased polymers, micro moulding, bioengineering and biomedical applications.