Proceedings of the
7th Graz Brain-Computer Interface Conference 2017 DOI: 10.3217/978-3-85125-533-1-92

HEADGEAR FOR MOBILE NEUROTECHNOLOGY: LOOKING INTO ALTERNATIVES FOR EEG AND NIRS PROBES

A. von Lühmann[1,2], S. Soekadar[3], K.-R. Müller[1,4], B. Blankertz[2]

[1] Machine Learning Group, Berlin Institute of Technology, Berlin, Germany
[2] Neurotechnology Group, Berlin Institute of Technology, Berlin, Germany
[3] Applied Neurotechnology Lab, University Hospital of Tübingen, Tübingen, Germany
[4] Dept. of Brain and Cognitive Engineering, Korea University, Seoul, Korea

E-mail: vonluehmann@bimos.tu-berlin.de

ABSTRACT: Brain-computer interfaces are now entering real-life environments. Particular hybrid systems using more than one input signal, e.g. electroencephalography (EEG) and functional near-infrared spectroscopy (fNIRS), offer a broad spectrum of applications in basic research and clinical neuroscience. Here, we provide an overview of recent EEG-electrode and fNIRS-optode approaches that aim to improve usability. We include our new multi-function clip-on design that allows the use of conventional gel-based ring electrodes with water. For EEG electrode approaches (conventional gel, solid gel, new custom water-based) we compared impedances and frequency response over multi-hour recordings. While the water-based solutions showed comparable performance in terms of signal quality, applicability and comfort, solid-gel electrodes on hairy skin required additional contact pressure. Overall, however, all tested EEG electrode types were well compatible with concurrent fNIRS recordings using a novel hybrid fNIRS/EEG headgear, paving the way for cognitive workload experiments under real-life conditions.

INTRODUCTION

In the past years, electroencephalography (EEG) and (continuous wave) functional near-infrared spectroscopy (NIRS) have significantly advanced towards higher miniaturization and mobility. While EEG allows assessing brain electric activity with high time- but low spatial precision, NIRS measures the metabolic (oxygen-dependent) activity of brain areas with high spatial but lower temporal precision. In general, this new generation of wearable hardware enables new approaches in many domains related to neuroscience and neurotechnology, such as in diagnostic medicine, cognitive science, psychology and brain-computer-interfaces (BCIs).

In non-invasive BCI, traditional approaches provide an active communication interface, e.g. for severely impaired or locked-in patients. Taking advantage of the recent trends in wearable instrumentation, BCI research increasingly focuses on domains beyond these applications [1] [2]. Amongst them are exoskeletons [3], [4], rehabilitation and mobility aids as well as adaptive neurotechnology research [5], [6] and neuroergonomics [7]: These BCIs aim to improve work environments, efficiency and security while advancing the understanding of brain function in everyday life scenarios. However, taking non-invasive neurotechnology further out of the lab and into everyday-life environments bears several multidisciplinary challenges.

I. *New instrumentation* needs to fulfill requirements w.r.t. size, power consumption, weight and cost while maintaining high acquisition performance in single wireless devices and body sensor networks
II. *Signal and analysis approaches* have to face lower signal to noise ratios (SNR), and increased influence of non-stationarities due to environmental influences and physiological artifacts, while further progressing with respect to robustness and overall accuracy / performance.
III. *Usability*: To achieve a wider adoption of neurotechnology outside of the laboratory, ergonomic aspects such as user-friendliness, preparation and set up times play a key role. Here, the interface between acquisition hardware and user is subject to research and developments. This includes headgear and fixation concepts as well as EEG electrodes and fNIRS optodes.

One way of tackling these challenges is by jointly using EEG and NIRS, for instance in *hybrid BCIs* [2], [8]. These bi- or multimodal systems exploit shared and complementary information in the acquired signals and use their respective strengths with respect to spatial and temporal resolution and artifacts to improve performance [9], [10]. Both technologies are relatively low cost and can be miniaturized and integrated in portable / wearable devices. We did this in our Mobile Modular Multimodal Biosignal Acquisition (M3BA) architecture [11], designed to tackle many of the challenges in mobile and out-of-the lab BCI. The M3BA modules are wireless miniaturized hardware for joint acquisition of EEG, fNIRS, electrocardiography (ECG) / electromyography (EMG) and accelerometer signals. We now integrated two modules into a new headset for mobile hybrid-BCI based experiments towards out of the lab scenarios. For headgear in mobile and life-like scenarios, optimization of usability and ergonomics of electrode

 DOI: 10.3217/978-3-85125-533-1-92

and optode solutions are crucial. Several non-traditional solutions have been presented in both research and industry in the past years [12] [13].

In this paper, we present results of our evaluation and work towards practical and user-friendly electrode/optode solutions for mobile EEG-NIRS based neurotechnology applications. We present a custom water-based open electrode design concept that enables the cost effective and easy upgrade of gel-based AgCl ring electrodes to water-based electrodes, allowing full integration into existing gel-based technology, such as the EasyCAP ring electrode products. We provide evaluation results of impedance measurements using different new electrode types and characterize the so far outstanding frequency response of solid-gel electrodes from [14] in a new epidermis-based experimental setup. Furthermore, we briefly discuss our developments in the field of spring-loaded and flexible NIRS optode holders.

MATERALS AND METHODS

Here, we categorize and summarize (without claim of completeness) existing solutions for EEG-electrodes and NIRS-optodes and our new approaches (see fig. 1, new approaches encircled in blue). We then present methods conducted for evaluation.

Traditional EEG electrodes

In EEG, a standard solution is the use of head caps made from stretch fabric with wet or dry electrodes at positions specified by the 10-20 (or 10-5) system. Conventional wet electrodes are usually based on AgCl material with an electrolyte (NaCl) based gel that also acts as a buffer, reducing motion and shift artifacts. Because a second person is required for the application of gel and hair cleaning after the measurements, usability of conventional wet gel electrodes is not optimal.

Dry electrodes are usually active (including built-in pre-ampliers), need no further preparation and rely on direct contact to skin. Different numbers and shapes of pins for skin-contact / hair penetration are available. They have higher impedances and thus show lower SNR and need a relatively high contact pressure that many users perceive as unpleasant over time. Furthermore, they are more prone to movement artifacts.

When there is no or little hair, sticky electrodes can be directly applied to the skin and provide good signal quality. Due to hair on the head, this type of electrodes is, however, usually not suitable for EEG.

With these electrode types, integration of BCI into everyday life and combination with wearable equipment seems not feasible. To achieve this, other solutions are more promising

Recent EEG electrode approaches

To perform EEG measurements in / around the ear, dry or wet electrodes in the ear [15] and sticky C-shaped electrodes around the ear [16] were proposed. For on-scalp measurements, many recent solutions focus on wet alternatives to sticky gel, such as non-adhesive new solid gel electrodes [14] and water-based electrodes. The latter combine sponges, pressed up wool

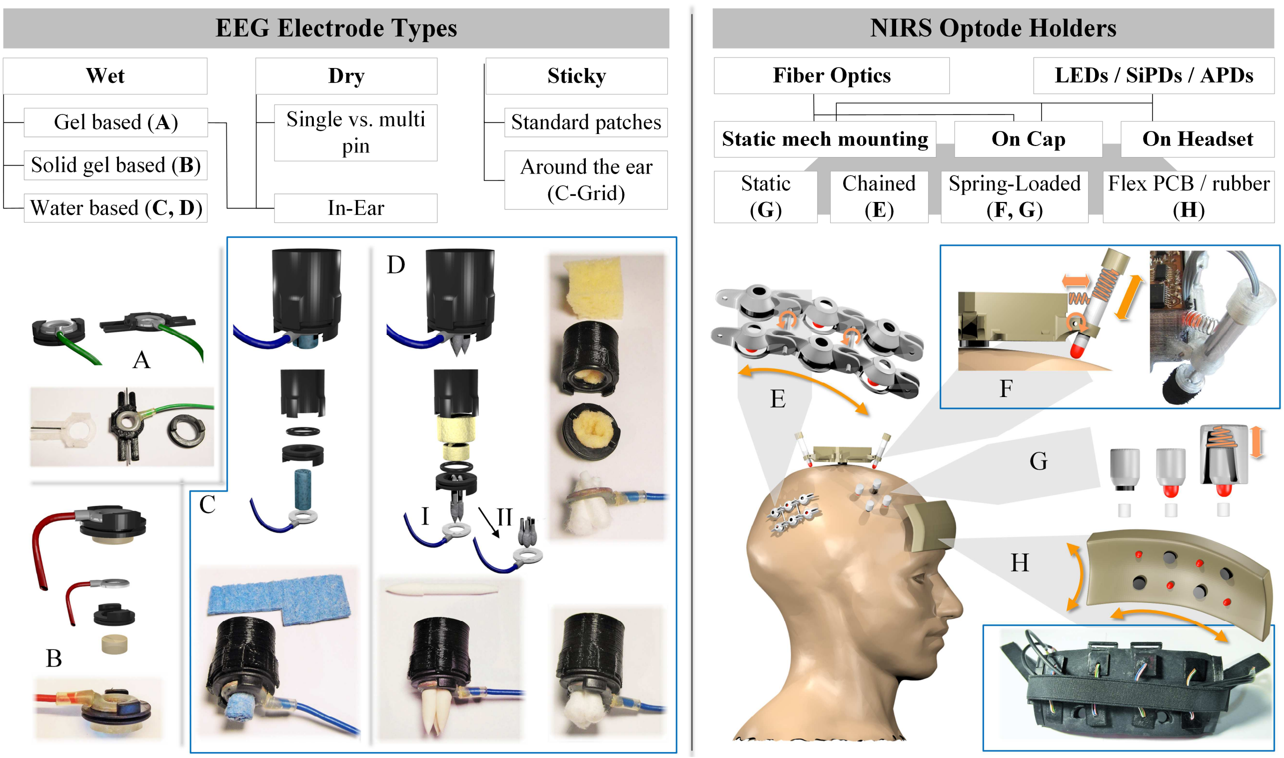

Figure 1: Strategies for (wet) EEG electrodes and NIRS optode attachment, our approaches encircled in blue. A: conventional gel ring electrode; B: ring electrode with solid-gel from [14], C/D: our new water-based adapter for EasyCAP ring electrodes. C: with rolled sponge/pressed cotton similar to [17], D: In a new approach with sponge and soft synthetic felt brush pen tips (I) or cotton bud heads (II). E: chain-link optode holders as in [21]; F: our spring-loaded openNIRS design [24]; G: conventional, as in [19]; H: flexible PCB (as in [22]) / rubber mat (our M3BA headset).

 DOI: 10.3217/978-3-85125-533-1-92

or rolled-up cotton [17] soaked in a NaCl dilution with AgCl electrodes and have been successfully applied in BCI applications [18].

Novel Developments

In a new water-based approach, we developed an easy-to-apply upgrade for EasyCAP ring-electrodes enabling a variety of applications. The electrodes can either be directly attached to a headset or to a 10-20 cap and allow traditional use with gel (fig. 1A), solid-gel (fig. 1B) and our custom water-based solutions:
Core of the invention is a 3D-printed clip-on reservoir for standard Easycap B10 series ring-electrode holders. The electrode is used upside-down, with its surface facing up towards the reservoir. The reservoir is clipped onto the holder with an O-ring seal. When used with cut and rolled sponge cloth (fig. 1C) stuck through the center of the electrode and a water based NaCl-shampoo solution, the resulting electrodes can be used similarly to [17] and TMSI water-electrode products. However, rolled cotton or sponge cloth creates a single contact surface that is more likely to be obstructed from direct skin contact by dense hair. Fig. 1D I) and II) show another new approach using our clip-on design. The reservoir is filled with sponge material soaked with the water based NaCl-shampoo solution. Either contact to the scalp is established with three synthetic felt brush-pen tips (fig. 1D I) or three cotton bud heads with hollow tubes (fig. 1D II). Similar to multi-pin dry electrodes, these soft pins can pass obstructing hair more easily. While being far more comfortable than solid dry metal pins and enabling direct (water-based) conduction, they offer only minimal surface for evaporation of water.

NIRS optode Approaches

In continuous wave NIRS, solutions depend on the optode architecture: many older instrument generations use rather heavy and bulky fiber optics to transfer light from the emission/detection hardware to the scalp. These are usually not suited for wearable equipment. Newer generations, few of which also support mobile applications, integrate LEDs and photo diodes into the optodes that are then attached directly to the scalp. They can be statically attached (fig. 1G) on stretch fabric caps similar to EEG [19], mechanical headsets or static mechanical mounting structures that do not allow the subject to move [20]. For wearable headset applications, chain link holders (fig 1.E) [21] for the whole head and flexible PCB / rubber mat pads for the forehead (fig 1H) [22] as well as forehead-covering headsets (e.g. [23]) were proposed.

Here, we implemented 4 NIRS emitters and detectors each, resulting in 10 channels (with 3.5cm optode distances), into a stretchy rubber mat piece with 3D printed mechanical holders for the user's forehead (fig 1H). The patch can easily be integrated into a headset and is the most convenient solution for non-haired regions, while at the same time blocking ambient light from the regions of interest.

Optical NIRS signals are prone to optode movements resulting in artifacts that are hard to distinguish from physiological changes in signals. Spring-loaded solutions (fig. 1F/G) can remedy this partly, but are mechanically more sophisticated. For headset-based optode attachment on hairy regions, we developed a double-spring-loaded optode solution (see fig. 1F). It ensures fast and easy perpendicular placement and accessibility of hair for preparation (for details please refer to the openNIRS project, [24]).

Experiment I: Evaluation of electrode impedances

To evaluate and compare the electrode-to-skin impedances of traditional gel (Abralyt 2000, EasyCAP), Mg and Ca based solid-gel and the new water-based electrode approaches, we conducted the following experiment. Two electrodes of each type (A, B, C, DI and DII) were applied to hairy regions of the head on 10-20 positions AF7/8, F1/2, C3/4, P5/6, AFz, Fpz and Oz on an EasyCAP. Impedances were measured using a commercially available signal amplifier (BrainAmp®, BrainProducts, Gilching, Germany), with a gel-based electrode reference on the left mastoid for A, C, DI, DII and a solid gel-based electrode reference on the right earlobe for B. Impedances were measured repeatedly every 30 min. over the course of 4 hours.

Experiment II: a) solid-gel impedances frequency response, b) solid-gel electrode impedances

Frequency responses and DC-characteristics of AgCl electrodes in a NaCl solution (gel- or water-based) are well known. Toyama et al. evaluated and compared their new solid-gel electrodes in several ways [14] and reported impedances and signals comparable to those from conventional paste-based recordings, but no frequency response. To enable a comparison of conventional gel and solid gel electrode frequency characteristics in the EEG-Spectrum over time, we now performed an experiment repeatedly measuring the

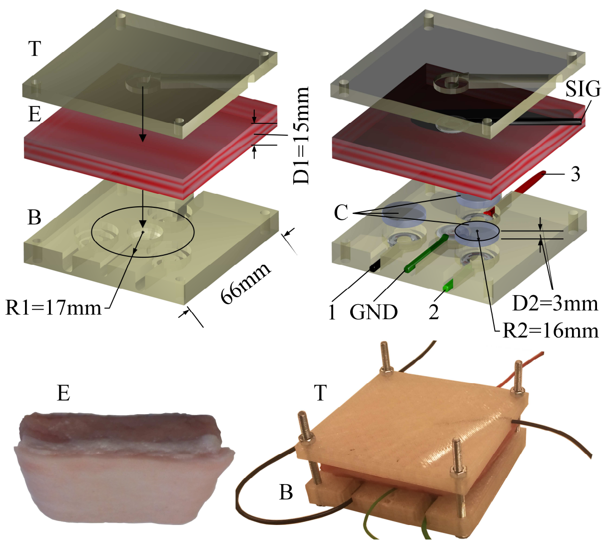

Figure 2: Custom 3D-printed holders, epidermis sample (E) with thickness D1, sine signal (SIG) and ground electrode (GND). Measurement electrodes (1, 2, 3) placed equidistantly (R1) to GND, to each other and to SIG; are covered by cylindrical conductive volumes (C) with defined radius (R2) and thickness (D2): Solid gel based on Mg (electrode 1), solid gel based on Ca (electrode 2), conventional electrolyte gel (electrode 3).

 DOI: 10.3217/978-3-85125-533-1-92

frequency responses of both the Mg/Ca based solid-gel and sticky paste (Abralyt 2000) together with conventional AgCl ring electrodes.

To enable a realistic, but controlled, close to real-life measurement scenario that includes interaction of the materials with skin/tissue over longer periods of time, we designed a custom 3D-printed holder. It allows precise positioning of the electrodes and electrolyte material on an epidermis-based phantom using a tissue sample of pork belly with 15 mm thickness and conventional AgCl EEG ring electrodes (see figure 2). Two 3D-printed holders (T, B) fixate the electrodes on the epidermis sample (E) at defined positions. An adjustable sine current is impressed into the tissue between a centered signal electrode (SIG) and a consumer load electrode (GND) that is connected to signal generator ground via a 150 Ω resistance. Signal measurement electrodes (1, 2, and 3) are placed equidistantly (R1= 17 mm) around the center GND electrode, furthermore equally distant from each other and from the signal (SIG) electrode. Between the measurement ring electrodes and the epidermis surface, cylindrical volumes with radius (R2) 16mm and height (D2) 3mm are filled with the conductive materials that are to be investigated: Mg based solid gel pads (electrode 1), Ca based solid gel pads (electrode 2) and standard abrasive electrolyte gel (EasyCAP Abralyt 2000) (electrode 3).

We performed three experiments at a constant room temperature of 23±1 °C, using three epidermis samples over the course of 4 hours. The impedance and frequency dependent attenuation of each channel (1-3) were measured within intervals of 30 minutes using a 100 mV_{pp} offset free sine signal in the range of 1-200 Hz with the following frequency steps: 1-50 Hz in 1 Hz steps, 50-200 Hz in 10 Hz steps.

Signals were generated with an Agilent Technology DSO-X 2014A device and acquired at all electrodes using a National Instruments USB-6003 data acquisition card. For each step in the frequency spectrum, 2 s were sampled with 1 kHz / 16 Bit resolution. From the datasets, average AC root mean square values and spectral power in the target frequency using FFT (Hamming window) were calculated for each frequency and each channel. Attenuation factors for the overall frequency response [dB] were calculated using the ratio of signal powers measured at the electrodes. The ADC's quantization limit and the maximum signal amplitude of 100 mV yielded a precision of the attenuation measurements of 0,026 dB. Tissue property influences that are assumed to be equal for all measurement channels as well as homogeneous changes over the course of the experiment (such as changes in moisture level and conductance), are minimized by pointwise subtraction: The differences $\Delta Att_{Mg/Ca}$ between frequency responses of the electrolyte gel $A_E(f)$ and of the solid gel electrodes ($A_{Mg}(f)$ and $A_{Ca}(f)$ respectively), depict deviations between the responses of both types.

$$\Delta Att_{Mg/Ca}\,[dB] = A_{Mg/Ca}(f) - A_E(f)\,[dB]. \quad (1)$$

Offsets of the average ΔAtt frequency responses, being all in the same order of magnitude, were then removed for better comparability, as they are based on constant tissue inhomogeneities between the electrode pairs.

RESULTS

Electrode impedances: Experiment I and IIb

The new designed reservoir-clip water electrode adapters (fig 1C and 1D I/II) were successfully applied in impedance measurement experiments and showed comparable EEG signal quality to that acquired with conventional gel. Figure 3 EI shows the impedances of all three water electrode types over the course of 4 h compared to gel. All three types showed impedances around 10 kΩ without additional preparation. Over the timecourse of the experiment, impedances dropped further below 10 kΩ.

The impedances of the solid gel electrodes showed poor performance on hairy regions in experiment I when no additional manual pressure was applied and were discarded from the set of data. Figure 3 EIIb shows the impedances of the Ca- and Mg-based solid gel electrodes measured in the epidermis probe

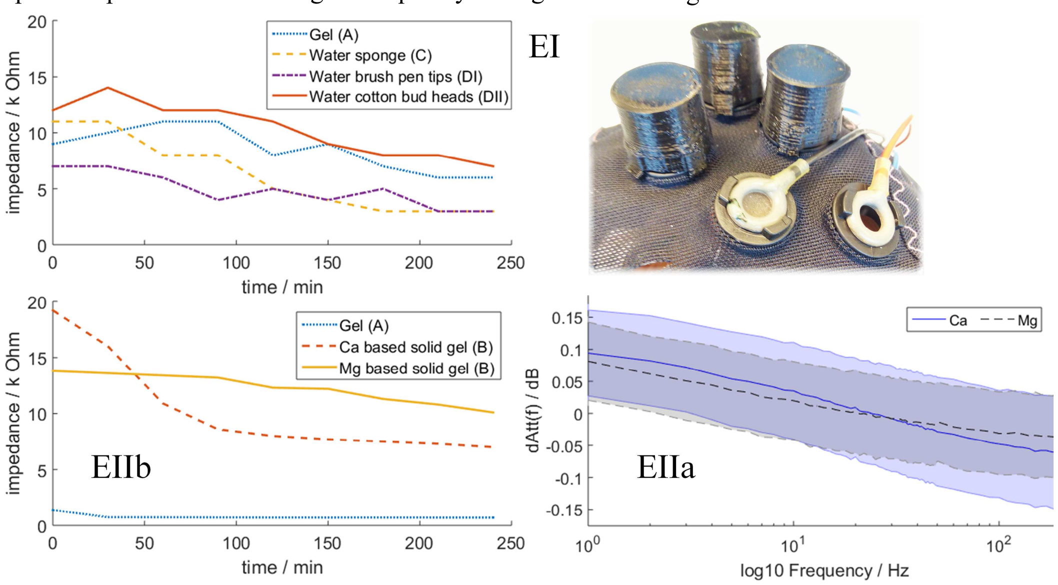

Figure 3: *EI*: Water vs. gel based electrode impedances and all 5 electrode types on EasyCap *EIIb*: Solid gel vs. gel based electrode impedances *EIIa*: grand average ΔAtt_{Mg} and ΔAtt_{Ca} over all three experiments and timepoints. Shaded bars: average standard deviation of $\Delta Att(f)$ over all timepoints.

 DOI: 10.3217/978-3-85125-533-1-92

Experiment IIb. These exemplify the solid-gel performance when applied directly on skin for longer time periods. Here, 14 kΩ and 19 kΩ were initially measured for the Mg-based and the Ca-based solid-gel respectively. Both impedances then further decreased significantly towards 8 kΩ (Ca) and 11 kΩ (Mg) over the timecourse of the experiment.

Solid gel frequency response: Experiment IIa

The average attenuation of the sine signal by the tissue was approx. -3 dB for all electrode types. Figure 3 EIIa shows the grand average delta frequency responses (all experiments and timepoints) for both solid-gel materials over the 4 hour experiments (blue solid and black dashed line). We observed a continuous change in the relative offsets of the delta frequency responses over the whole time-course. The grand average delta responses were roughly equal to the single responses measured at the midterm of the experiments. To depict the offset-shift of delta frequency responses relative to midterm over time, shaded bars show the 3-experiment average of the delta responses' standard deviation using the attenuation factors of all measured time points for a respective frequency bin. The measurements show negligible differences between the time-average frequency response of the electrolyte gel and both solid-gel materials over the range of 1-200 Hz. Over all time points, the delta frequency responses of both materials deviate less than ±0.06 dB from the average, which is only a little bit more than twice the measurements' precision bound of 0.026dB. Over the whole time-course, the maximum difference between the reference electrolyte's frequency response and Ca- and Mg- based solid gel responses was less than 0.31 dB and 0.24 dB, respectively.

DISCUSSION

All new water-based approaches (cotton bud heads, sponge cloth and brush pen tips) competed well with conventional gel electrodes and none excelled the others with respect to impedances and signal quality. While the new water-based types were easy to set up and comfortable to wear, reduction in their preparation time was not noteworthy. However, depending on the user's residual functional capability, they can decrease or obviate the need for support by a second person.

The solid-gel electrodes did not perform well on hairy regions without additional (manual) pressure. In the epidermis-based experiments however, they showed coinciding impedance results (range and time-courses) with those measured by Toyama et al. [14]. While the preparation of solid-gel electrodes required the least amount of time, the design and pressure of an optimized headset seems necessary. The frequency response evaluation results indicate that the time-averaged solid gel frequency responses deviate less than 0.155 dB (1.8%, Ca) and 0.11 dB (1.3%, Mg) from the conventional electrolyte over the range of 1-200Hz. While the attenuation is constant over all frequencies, the time courses of the attenuation offsets showed a total change of <0.31 dB for Ca and <0.24 dB for Mg based solid gel over the course of 4 hours. These offsets over time are likely due to the decline of solid gel electrode impedance over time in the experiment.

The set-up was designed symmetric and with equidistant electrode positions to enable a maximum comparability between the three measurement channels. However, despite the careful selection of tissue samples with minimal natural inhomogeneities, the exact impact of these tissue inhomogeneities on attenuation differences in the measurement channels cannot be determined completely. Thus, the acquired data can only be used for pessimistic upper bound estimates of the differences in the material's frequency responses.

However, we consider these upper bound estimates to be a good indicator that the solid-gel electrodes can be used for biopotential/EEG measurements leading to results comparable with conventional electrolyte gel.

CONCLUSION & OUTLOOK

Fusing the results of our work on hybrid EEG-NIRS acquisition and electrode/optode technology allowed us to design a new stand-alone hybrid EEG-fNIRS headset utilizing two M3BA modules and one rechargeable LiPo battery (up to 10 hours continuous acquisition), resulting in 9 EEG, 10 NIRS and 3 EMG/ECG channels and two 3D accelerometer signals (see figure 4). The headset weighs 150 g and can be worn on top of an additional EEG cap to enable change in number and standardized positions of electrodes, if desired. It enables the use of all electrode and optode solutions discussed in this paper and exemplifies one of several

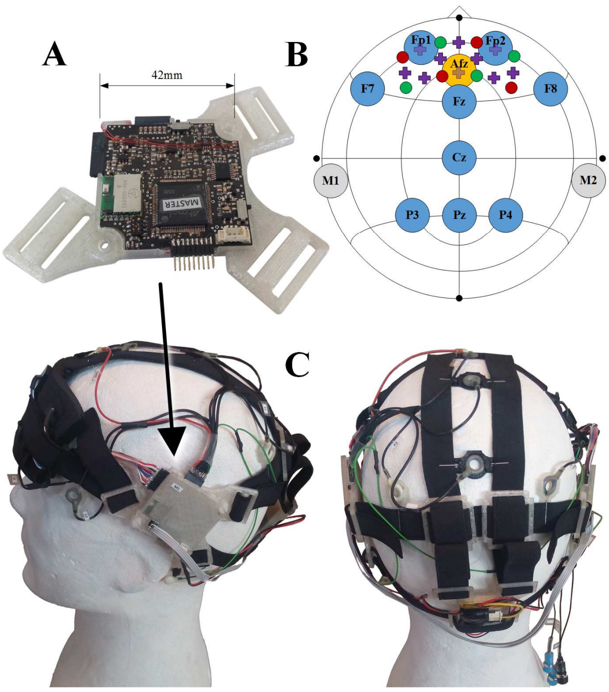

Figure 4: New hybrid EEG-NIRS M3BA headset. A: M3BA module in custom holder. B: channel placement. Blue: 10-20 EEG electrode positions; orange: GND; grey: linked mastoid reference; red/green: NIRS emitters/detectors; purple: NIRS channels. C: Headset.

 DOI: 10.3217/978-3-85125-533-1-92

alternatives for new mobile experiments. In this layout, the headset's channel placement was chosen to enable acquisition of frontal/parietal alpha/theta EEG power and metabolism, frontal asymmetries (in both the EEG spectrum and NIRS oxygenation signal), Error Potentials and Event Related Potentials. We will soon apply it in mobile cognitive workload detection experiments that are currently in preparation.

REFERENCES

[1] B. Blankertz, L. Acqualagna, S. Dähne and others, "The Berlin Brain-Computer Interface: Progress Beyond Communication and Control," *Frontiers in Neuroscience,* vol. 10, 21 November 2016.

[2] G. Müller-Putz, R. Leeb, M. Tangermann and others, "Towards Noninvasive Hybrid Brain-Computer Interfaces: Framework, Practice, Clinical Application, and Beyond," *Proceedings of the IEEE,* vol. 103, no. 6, pp. 926-943, 2015.

[3] N.-S. Kwak, K.-R. Müller and S.-W. Lee, "A lower limb exoskeleton control system based on steady state visual evoked potentials," *Journal of Neural Engineering,* vol. 12, no. 5, p. 056009, 2015.

[4] S. Soekadar, M. Witkowski and C. Gómez, "Hybrid EEG/EOG-based brain/neural hand exoskeleton restores fully independent daily living activities after quadriplegia," *Science Robotics,* vol. 1, no. 1, p. eaag3296, 2016.

[5] K.-R. Müller, M. Tangermann, G. Dornhege and others, "Machine learning for real-time single-trial EEG-analysis: From brain-computer interfacing to mental state monitoring," *Journal of neuroscience methods,* vol. 167, no. 1, pp. 82-90, 2008.

[6] T. O. Zander and C. Kothe, "Towards passive brain-computer interfaces: applying brain-computer interface technology to human-machine systems in general," *Journal of neural engineering,* vol. 8, no. 2, p. 025005, 2011.

[7] R. Parasuraman, "Neuroergonomics: Brain, Cognition, and Performance at Work," *Current Directions in Psychological Science,* vol. 20, no. 3, pp. 181-186, 2011.

[8] G. Pfurtscheller, B. Allison, G. Bauernfeind and others, "The hybrid BCI," *Frontiers in Neuroscience,* vol. 4, no. 3, 2010.

[9] S. Fazli, J. Mehnert, J. Steinbrink and others, "Enhanced performance by a hybrid NIRS-EEG brain computer interface," *Neuroimage,* vol. 59, no. 1, pp. 519 - 529, 2012.

[10] S. Dähne, F. Biessmann, W. Samek and others, "Multivariate Machine Learning Methods for Fusing Multimodal Functional Neuroimaging Data," *Proceedings of the IEEE,* vol. 103, no. 9, pp. 1507-1530, 2015.

[11] A. von Lühmann, H. Wabnitz, T. Sander and others, "M3BA: A Mobile, Modular, Multimodal Biosignal Acquisition architecture for miniaturized EEG-NIRS based hybrid BCI and monitoring," *IEEE Trans. on Biomedical Engineering,* vol. in print, 2016.

[12] C. Grozea, C. Voinescu and S. Fazli, "Bristle-sensors - low-cost flexible passive dry EEG electrodes for neurofeedback and BCI applications," *Journal of neural engineering,* vol. 8, no. 2, p. 025008, 2011.

[13] F. Popescu, S. Fazli, Y. Badower and others, "Single Trial Classification of Motor Imagination Using 6 Dry EEG Electrodes," *PloS one,* vol. 2, no. 7, p. e637, 2007.

[14] S. Toyama, K. Takano and K. Kansaku, "A non-adhesive solid-gel electrode for a non-invasive brain-machine interface," *Frontiers in Neurology ,* vol. 3, 2012.

[15] D. Looney, P. Kidmose, C. Park and others, "The In-the-Ear Recording Concept: User-Centered and Wearable Brain Monitoring," *IEEE Pulse,* vol. 3, no. 6, pp. 32-42, 2012.

[16] S. Debener, R. Emkes, M. De Vos and others, "Unobtrusive ambulatory EEG using a smartphone and flexible printed electrodes around the ear," *Scientific reports,* vol. 5, p. 16743, 2015.

[17] I. Volosyak, D. Valbuena, T. Malechka and others, "Brain–computer interface using water-based electrodes," *Journal of Neural Engineering,* vol. 7, no. 6, p. 066007, 2010.

[18] V. Mihajlovic, G. Garcia Molina und J. Peuscher, „To what extent can dry and water-based EEG electrodes replace conductive gel ones?: A Steady State Visual Evoked Potential Brain-computer Interface Case Study," in *ICBE 2011: Int. Conf. on Biomed. Eng.*, Venice, 2011.

[19] S. K. Piper, A. Krueger, S. P. Koch and others, "A wearable multi-channel fNIRS system for brain imaging in freely moving subjects," *NeuroImage,* vol. 85, pp. 64-71, 2014.

[20] S. Coyle, T. Ward and C. Markham, "Brain-computer interface using a simplified functional near-infrared spectroscopy system.," *Journal of Neural Engineering,* vol. 4, pp. 219-226, 2007.

[21] J. Safaie, R. Grebe, H. Moghaddam and others, "Toward a fully integrated wireless wearable eeg-nirs bimodal acquisition system," *Journal of Neural Engineering,* vol. 10, no. 5, p. 056001, 2013.

[22] K. Izzetoglu, S. Bunce, M. Izzetoglu and others, "Functional near-infrared neuroimaging," *IEMBS '04. 26th Ann. Int. Conf. of the IEEE,* vol. 2, pp. 5333-5336, 2004.

[23] M. Kiguchi, H. Atsumori, I. Fukasaku and others, "Note: Wearable near-infrared spectroscopy imager for haired region," *Review of Scientific Instruments,* vol. 83, p. 056101, 2012.

[24] A. von Lühmann, C. Herff, D. Heger and others, "Towards a wireless open source instrument: functional Near-Infrared Spectroscopy in mobile neuroergonomics and BCI applications," *Frontiers in Human Neuroscience,* vol. 9, no. 617, 2015.