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Bristle-sensors—low-cost flexible passive dry EEG electrodes for neurofeedback and BCI applications

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Abstract

In this paper, we present a new, low-cost dry electrode for EEG that is made of flexible metal-coated polymer bristles. We examine various standard EEG paradigms, such as capturing occipital alpha rhythms, testing for event-related potentials in an auditory oddball paradigm and performing a sensory motor rhythm-based event-related (de-) synchronization paradigm to validate the performance of the novel electrodes in terms of signal quality. Our findings suggest that the dry electrodes that we developed result in high-quality EEG recordings and are thus suitable for a wide range of EEG studies and BCI applications. Furthermore, due to the flexibility of the novel electrodes, greater comfort is achieved in some subjects, this being essential for long-term use.

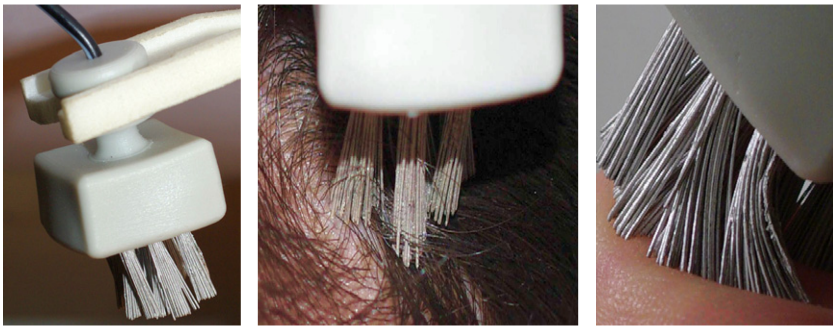
1. Background and objective

While there has been a recent surge in dry electrode technology, with many groups starting research in this domain [1–3], dry electrodes were proposed back in the early 1990s [4, 5], and early pioneering work on capacitive electrodes had already begun in the early 1970s [6]. The miniaturization of EEG equipment [7], as well as the wearability and convenience of novel EEG systems, will be a vital factor in determining whether EEG-based related BCI technology will be accepted by the wider community and thus gain wide-spread use. Being able to measure high-quality EEG signals through hair, while at the same time not posing any health risks, is also very important.

Classic gel-based electrodes present a number of inconveniences that have prevented this spread so far: the setup of an EEG cap is time-consuming, and also the gel is wet; it may dry up and require washing of the hair after use. In addition, the drying up gel can lead to varying impedances and the need for periodical recalibration, making long-term monitoring more difficult.

State-of-the-art dry electrodes for EEG can be grouped into two categories: ones that on purpose penetrate the outermost layer of the skin (*stratum corneum*) in order to eliminate its impedance from the recording chain, and the other, nonskin-invasive electrodes. The second category can be split further between contact electrodes and capacitive coupling electrodes.

Two recently proposed solutions of the skin-invasive category use microtip electrodes, coated with iridium oxide [8] and carbon nanotubes [9] (Enobio). As one may expect, both methods lead to high signal quality and are from this perspective well suited to EEG recording. However, the possibility of the microtips breaking off while partly still inside the skin cannot be completely ruled out [8]. For the case of carbon nanotubes, the authors also mention these safety concerns, for which reason they have performed only limited tests with a single subject [9]. The infection risks of electrodes that penetrate tissue is examined in [10], concluding that sterilization (not only disinfection) of the electrodes is needed to rule out the transfer of infectious diseases, such as human immunodeficiency virus (HIV), hepatitis C virus (HCV) and Creutzfeldt–Jakob Disease (CJD). For the latter, even sterilization is mentioned as not being adequate to destroy the prion that transmits CJD. These risks speak against the



(a) Prototype (b) Through hair (c) Flexing

Figure 1. The silver-coated polymer bristles prototype: (a) prototype, (b) through hair and (c) flexing.

use of skin-invasive dry EEG electrodes other than disposable versions, irrespective of production cost.

While capacitive electrodes have been successful in the extraction of steady-state visual evoked potentials [11], we are not aware of a publication where neurophysiological responses, such as event-related (de-)synchronization (ERD/ERS), have been shown. Furthermore, it has been reported that capacitive electrodes tend to be strongly affected by motion and muscular artifacts, and also exhibit long recovery times after such artifacts occur [3, 12].

To conclude, we may state that while there has been a lot of progress recently, there are still a number of problems in the domain of dry electrode research that need to be addressed. In light of this, it is not surprising that even wet electrodes are still actively being developed [7, 13].

Interesting direct-contact nonskin-invasive dry electrodes are arrays of metallic pins [1, 14, 15] that have the advantage of being able to go through the hair of most subjects (two subjects out of seven reported in [1] as excluded from the experiment on these grounds).

The critical issue with contact nonskin-invasive electrodes is the pressure that has to be applied to get a low impedance of the electrode to skin contact (or equivalently a high admittance). The effect on the pressure on the impedance of the skin-to-electrode contact has already been established, e.g. in [16], where it is stated that 'Since the admittance is largely influenced by the contact pressure for a dry electrode, the pressure should be adjustable.' There the admittance was found to increase almost linearly with the pressure over the base value, for pressure within the range tested 0–500 g cm^{-2}. These results are in good agreement with our empirical observations with the electrodes reported in this paper and also with the previous ones developed at our lab and reported earlier [1, 17, 18].

What impedances are actually required for accurate EEG acquisition? A previous study [10] has shown that impedances of up to 200 kΩ still allow for accurate EEG signal acquisition (approximately 1% error). The need for low impedance contacts on the skin can in part be alleviated by the high input impedance of the amplifiers, shielding of the electrode wires and use of local impedance adapters (active electrodes). Still, even with local impedance adapters, the long-term pressure applied through array-of-pins electrodes can be unpleasant, as found in [14], where the author reports that 'the thin pin arrays can hurt your skin if you use them (for) too long'.

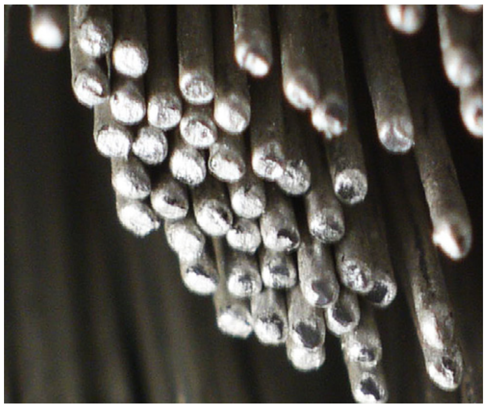


Figure 2. Macro-photograph showing the way the coating on the tips of the bristles is wearing out after months of use. The tips looking dark in this photography are no longer properly coated.

Here we propose a new dry EEG passive electrode (i.e. without signal amplification and impedance adapter on site) that improves existing pin-based nonskin-invasive dry EEG electrodes [1, 14, 15], in that it reduces the reported discomfort by distributing the pressure on the skin of the scalp more uniformly and more flexibly. The novelty consists in using flexible conductive bristles instead of pins. These bristles are made of silver-coated polymer in the reported prototype. With the proposed electrodes it is not only possible to reduce discomfort, a highly important aspect of useability, but also to increase the reliability of the contact, resulting in good signal quality.

2. Materials and methods

2.1. The prototype electrode

2.1.1. Description and mounting. The prototype, shown in figure 1(a), is soft because it has been obtained by coating thin polymer bristles with silver particles. The coating was accomplished by painting the bristles with a silver-based conductive ink. The coating is resistant to flexing the bristles, but it tends to wear out after multiple uses, especially on the tips of the bristles, as can be seen in figure 2. The bristles almost flex as easily as bristles in tooth-brushes, which were used as a source of the building material. Since the uncoated bristles were originally designed for contact with gums, which are of a far more sensitive skin-type than the *stratum corneum*, it is ensured that they do not penetrate the skin of the head and thus are easily accepted by the users. Since the cost of the electrodes is very low, they could easily be replaced and it would therefore be feasible to treat them as disposable electrodes if desired, allowing a hygienic and risk-free method to proceed for reasons discussed in the previous section.

The physical dimensions are 12 mm x 12 mm and the bristles are 10 mm long. The electrodes are mounted on a Speed Cap mount [19]. The Speed Cap is a patent pending mount developed in our lab, composed of one circular frame and several arms with positions adjustable on this frame. The electrodes are fixed at the end of the arms. The latest version of the Speed Cap offers the possibility of adjusting the position of the electrodes and the pressure towards the scalp, individually or in groups of three (when at the end of an arm a star configuration of three electrodes is mounted, instead of a single electrode). For the experiments reported here with the prototype, we have used an older version of the Speed Cap, without the screws used for adjusting the pressure; therefore, the pressure was only adjusted by moving the whole cap up or down slightly when fixing it on the head. The version we used has one spring at each arm so as to sustain stable contact. By compressing this spring more, the pressure increases according to Hooke's law. We adjusted the cap position until both subjects were still comfortable wearing it and the EEG signal monitored a reduction in the level of artifacts, which we know is correlated with acceptable impedance of the electrode-to-scalp contacts.

2.1.2. *Electrical characterization.*

Measurement of the electrode-to-skin impedance. To measure the electrode-to-skin impedance, a simple dc circuit has been created, powered by a low-voltage (5 V) pile. Two identical electrodes are put in contact with the scalp's skin and connected in series with the pile and the dc current measurement device. Using Ohm's law, the real component of the impedance was estimated, on the assumption that the two electrodes have identical impedance to skin and that the inner-body part of the circuit has negligible impedance. The estimation was further validated by replacing one of the dry electrodes with a gel-based electrode previously measured and repeating the measurement.

The impedance was measured at dc instead of e.g. 10 Hz (roughly the frequency of the alpha rhythms) in order to get an upper bound over the whole frequency range of interest. According to experimental research on the impedance of the skin, this impedance decreases with frequency [20], so what we obtain is an estimation of an upper bound of the impedance over the whole frequency range interesting for EEG (from dc to at least 600 Hz).

The half-cell potential of the electrode can interfere with our measurement, by making the effective voltage differ from the applied one with up to the sum of the half-cell potentials of the electrodes, but only when the electrodes are different or the temperatures at the electrodes differ. For reference, we mention here the absolute value of the standard reduction potentials of silver (Ag), $E_{\mathrm{Ag}} = 0.7996$ V, and silver chloride (AgCl), $E_{\mathrm{AgCl}} = 0.222\,33$ V [21].

Impedance. The electrode-to-skin impedance with freshly coated electrodes was typically 80 kΩ for low-to-medium, unobtrusive pressure on the scalp through hair (specific contact resistivity typically 11.5 $\Omega\mathrm{m}^2$ computed for the total surface of the electrode's side facing the scalp, $144 \times 10^{-6}\mathrm{m}^2$). After 10 months of wear, the same level of pressure as subjectively estimated by the users could only produce impedances in the range 150–200 kΩ (specific contact resistivity from 21.6 to 28.8 $\Omega\mathrm{m}^2$).

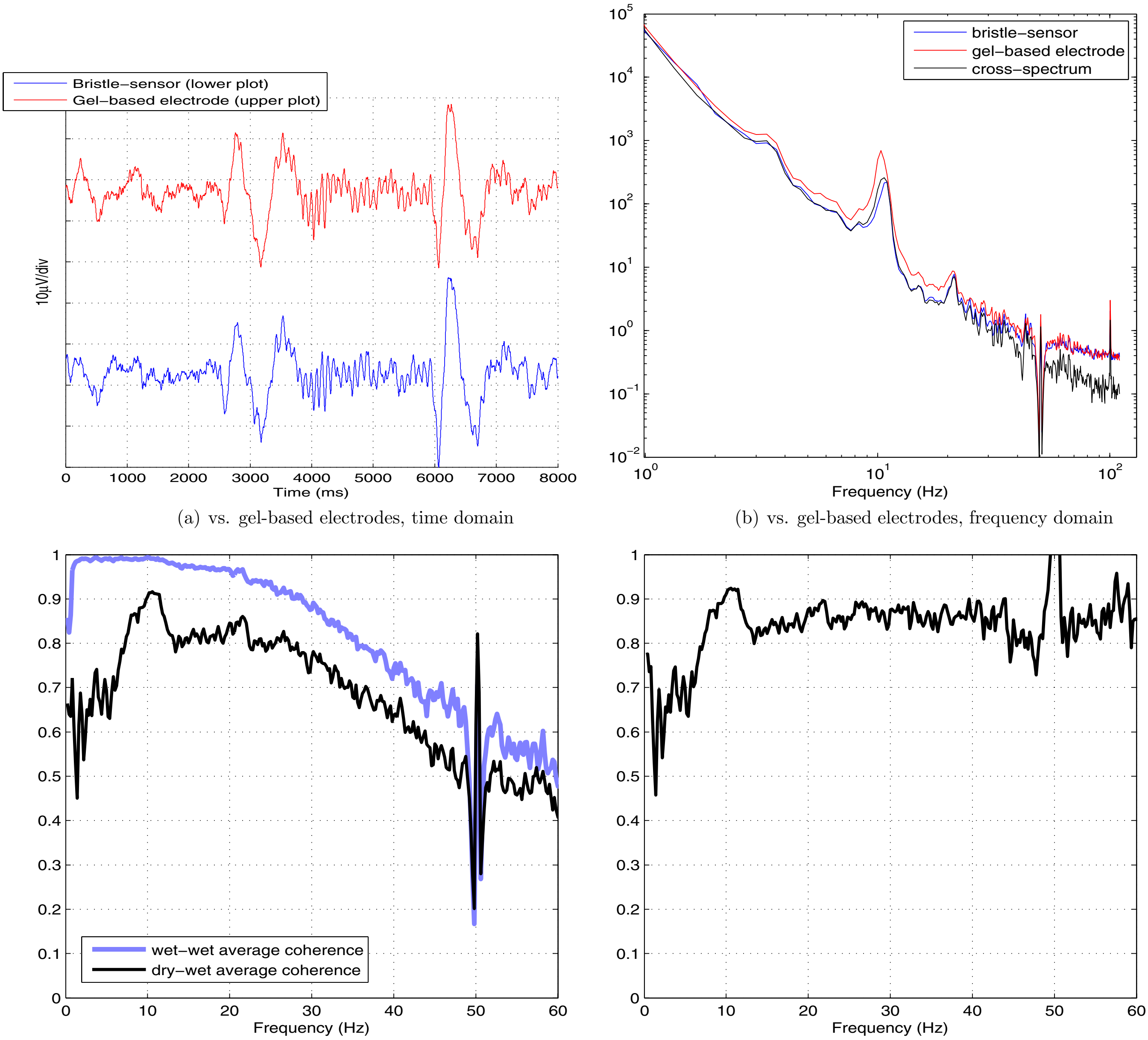
2.1.3. *Direct comparison with the gel-based electrodes.*

In order to validate the electrode, we performed some of the experiments described in section 2.2 in a setup where in close proximity to the dry electrode at P3 two other gel-based electrodes were placed, such that the distance between the centres of each two electrodes in the group was approximatively 2 cm. This enabled us to perform simultaneous recordings with gel-based and dry electrodes at very close locations, for a variety of tasks. The gel used was Abralyt 2000 (Easycap, Germany). When using the gel-based electrodes, the dry reference was replaced by a gel-based electrode on the nose.

A typical sample of time domain simultaneous recordings are shown in figure 3(a). The alignment of the spectra is shown in figure 3(b). In addition to the spectra, we have also computed the cross-spectrum on 5 s long moving windows, overlapping 1 s, using a Hanning window function (see [22] for the definitions of cross-spectrum, coherency and coherence). This is useful in that it offers further evidence for the similarity of the dry- and wet-channel signals, beyond the similarity of the shapes of the EEG spectra.

One can note that the main peak in the EEG at about 10 Hz and the secondary peak at about 20 Hz are both recorded fine (roughly the same signal-to-noise ratio) with the bristle-sensor as well.

In some papers, the overall correlation between the signals, which stem from wet and dry electrodes, is offered as proof that the new electrodes are providing almost the same signal quality as compared to standard gel-based electrodes [3, 15]. However, this measure can be easily influenced by one or more dominating subsignals, be it the 50/60 Hz, occipital alpha rhythm, EOG or motion artifacts. This does not capture how accurate the novel signal is as a function of frequency and thus determining which EEG studies and BCI applications are possible with the novel technology and which are not. We have therefore performed a more detailed analysis; using the spectra and the cross-spectrum, we have computed the coherence at each frequency. The averaged results for seven experiments of two subjects (AA, AC) are shown in figure 3(c). The average coherence of the bristle-sensor/gel-based pair is above 80% of the average coherence of the two gel-based electrodes, from



(a) vs. gel-based electrodes, time domain

(b) vs. gel-based electrodes, frequency domain

(c) average coherence dry-wet and wet-wet

(d) dry-wet coherence normalized by wet-wet coherence

Figure 3. Signal quality assessed by direct comparison with simultaneously recorded signal with gel-based electrodes. (a) Sample time domain signal accompanied by signal from a gel-based electrode on a neighbouring location, after band-pass filtering 1–45 Hz; alpha rhythm is visible from $t = 4000$ ms to $t = 6000$ ms. (b) Spectra and cross-spectrum for prototype and a gel-based electrode on a neighbouring location, after rejection of 50 Hz. (c) Average over the experiments of two subjects of the coherence between the dry and the wet electrodes signals; for comparison, the average coherence is computed for a pair of wet electrodes. (d) Ratio of the average coherence dry–wet and the average coherence wet–wet.

7 to 44 Hz (figure 3(d)). In the frequency range of the alpha waves (around 10 Hz), the dry-wet average coherence reaches in excess of 90% of the wet–wet average coherence.

2.2. Experiments

Eight volunteers, all working in our lab, gave their informal consent to participate in testing the new electrode, by means of a number of experiments. They were informed about the nature of the sensor and they were asked to stop the experiment in case of any discomfort. For two of them, we only performed some very basic checks, as we have noticed that the early prototype of Speed Cap did not fit well; therefore, there was no way to increase the pressure on their scalp (the electrodes were in part floating in the air) and thus there was no way to achieve a workable contact impedance, below 300 kΩ. One other volunteer agreed only to briefly test the prototype electrodes for the comfort survey, but not to participate in the experiments. The volunteers were not paid for their participation. They are referred here with randomized initials. The five subjects who took part in the experiments were four men and one women,

Table 1. Sessions and results.

Subject/age of electrode (months)	alpha classification accuracy (%)	N100 (*t*-test *p*-value)	P300 (*t*-test *p*-value)	BCI training /feedback accuracy	Comments
AA/1	90	10^{-12}	10^{-6}		
AA/10		10^{-7}	10^{-8}	87.5 ± 14.9% /87.2%	Two gel-based electrodes near P3; gel-based reference on the nose; BCI with real executed movements, leveraging the similarity between the cortical activation for executed and for imagined movements [25]
AB/1	75.8	10^{-10}	10^{-2}		
AC/1	83	10^{-4}			
AC/10		10^{-3}	10^{-4}		Two gel-based electrodes near P3; gel-based reference on the nose
AC/1				92.5 ± 14.9% /71%	After recoating. Real executed movements
AD/7				78.8 ± 13.6% /79%	
AH/1	68.3	10^{-2}			After recoating

with an average age of 32.6 ± 3.4 years, good hearing and good vision.

A summary of the sessions carried out is given in table 1 and below we give a detailed description of each experiment type that we performed. Electrode positions were placed and are indicated with labels according to the international 10–20 system for EEG electrode placement.

The experiments we performed were occipital alpha rhythm detection (alpha), N100 auditory evoked potential (N100 AEP), auditory evoked P300 event-related potentials (P300), and sensory-motor rhythm (SMR)-based BCI motor imagination brain–computer interface tests (BCI).

A multi-channel EEG amplifier (BrainAmp DC by Brain Products, Germany) was used to acquire the EEG signal in all experiments mentioned. The amplifier was set to 10 MΩ input impedance, and thus the implicit voltage divider ratio was in the range of 0.98–1, when the electrode-to-skin impedance was under 200 kΩ and the electrode was connected directly to the amplifier. The sampling rate was 1000 Hz. A hardware low-pass filter with a cutting frequency of 250 Hz, integrated into the EEG amplifier, was enabled.

2.2.1. Alpha rhythm. For the detection of the alpha rhythm, we recorded the EEG signal at the central occipital channel *Oz*. Subjects were instructed to fixate a cross for 30 s and then to keep their eyes closed, also for 30 s. The separability of the eyes open and eyes closed conditions was then estimated offline with a leave-one-out cross-validation of linear classification on spectra densities of 1 s long segments. For classification, spectral power in 2 Hz wide bins, from 8 to 14 Hz, was computed with 1024 ms long moving windows with a 960 ms overlap, using a Hanning window function. We used linear discriminant analysis (LDA) for classification. The contrast between the spectra corresponding to the two conditions, averaged over all subjects with available data, is displayed in figure 4.

2.2.2. N100 AEP. The experimental procedure for the N100 followed a standard paradigm [17]: the N100 auditory evoked potential stimuli were 45 ms long, with a frequency of 1.3 kHz, an amplitude of 90 dB and spaced with random intervals of 1–1.5 s, in order to prevent subject habituation. The EEG signal was recorded at *Cz*, with an identical dry reference electrode at *Fz*.

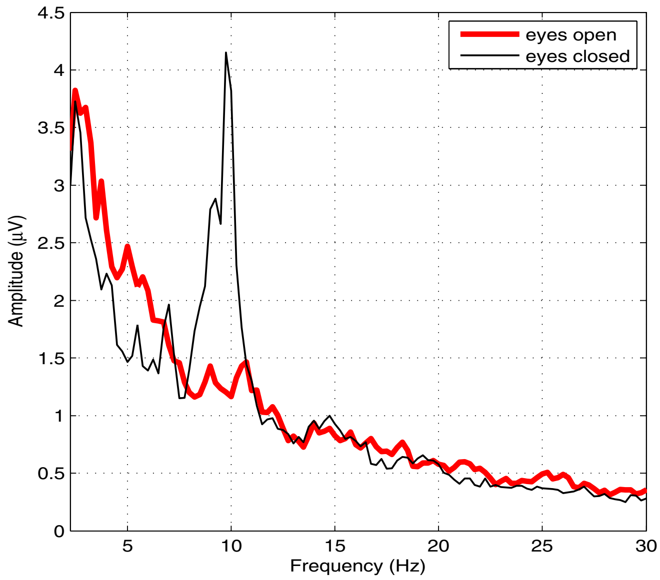


Figure 4. Spectra of the EEG signal recorded with the prototype for the eyes open/eyes closed conditions, averaged over all subjects with available data. This shows a peak at 10 Hz for the eyes closed condition.

The analysis of the N100 signals was band-pass filtering 2–16 Hz (Butterworth filter of order 6), followed by averaging of stimulus-aligned segments, after trial-wise baseline referencing with a window of 20 ms before stimulus presentation.

The statistical significance of the effect was computed with one sample *t*-tests for every sampled timepoint from −20 to 650 ms and cumulative for each new presentation.

The outcome for one experiment is shown in figure 5. The upper part shows the average voltage and the standard

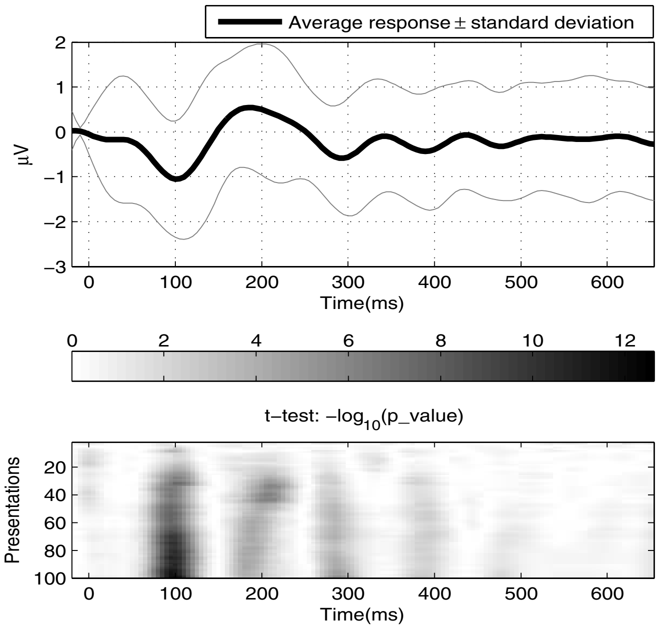


Figure 5. N100 effect and statistical significance as a function of the number of presentations.

deviation. Note that the very low standard deviation at approximately −10 ms is due to the referencing. The lower part of the plot shows the evolution of the statistical significance.

The outcomes for all subjects are summarized in table 1.

2.2.3. P300 ERP. The experimental procedure for the P300 tests followed [18]: auditory P300 was elicited by 45 ms long, 60 dB tones produced at random intervals of 1–1.5 s. 700 Hz for the normal stimuli and 1.3 kHz for the oddball stimuli were used and the latter tones occurred with a relative frequency of 1/5. The data were recorded with a dry reference electrode at *Fz*, and two dry electrodes placed at *Cz* and *P1*—all three electrodes were identical and were placed on a 'Speed Cap' mount.

The analysis of the P300 signals followed the following steps: band-pass filtering 2–16 Hz (Butterworth filter of order 6), Jade ICA [23], followed by condition-wise averaging of stimulus-aligned segments, referenced by a window of 40 ms before stimulus presentation.

The statistical significance of the difference between the two conditions was computed with two-sample t-tests for every sampled timepoint from −40 to 650 ms and cumulative for each new presentation of the oddball stimulus. The null hypothesis is that the two subpopulations of values (corresponding to the normal and the oddball stimuli) have the same mean at the same time after stimulus presentation (e.g. 300 ms), and the alternative is that the means are different. The test can be repeated when more evidence is gathered, and this is done here: the test is repeated after every new presentation of the oddball stimulus. Thus, for each time point from −40 to 650 ms and each presentation of an oddball stimulus, a new p-value is computed, taking into account all responses to stimuli presented so far (the lower part of figure 6). In table 1, we report the minimum p-values at around 300 ms for each P300 experiment. For the given values, the entire set of presentations is taken into account and the result can be interpreted as an indication of the presence and strength of the P300 effect.

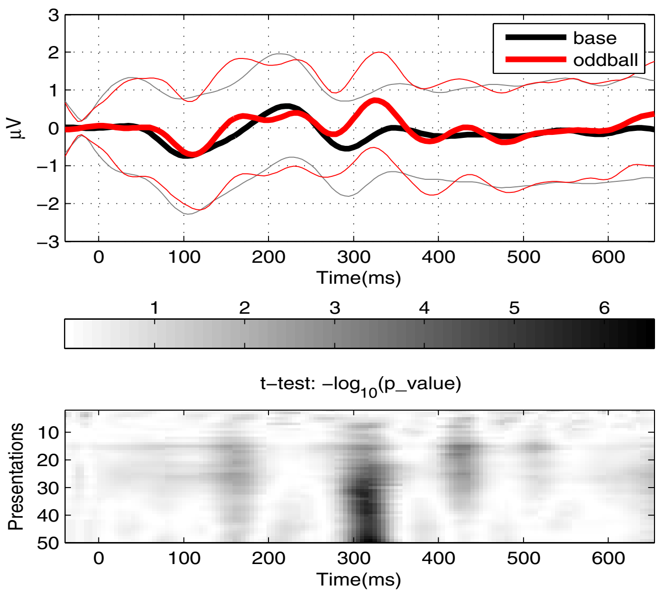


Figure 6. P300 effect and statistical significance as a function of the number of presentations.

The outcome for one subject is displayed in figure 6: the 50 trials (red, oddball stimuli) and 250 trials (blue, normal stimuli) averaged responses are given with standard deviation, as well as the evolution of the statistical significance.

2.2.4. BCI. For the BCI sessions, we used a configuration of five electrodes: one frontal reference at *Fz*, three electrodes covering the motor cortex of the left hand (*C4, FC2, CP2*) and one electrode at *P3*. This configuration of electrodes targeted two classes: left-hand movement imagination and the 'relax' state. All electrodes were placed on a Speed Cap mount. Apart from the different electrodes used, the procedure described in [17] was followed. The software system used was the Berlin BrainComputer Interface—BBCI [24]. The two classes used in the BCI session for one subject were imagination of the left-hand movement and that of the right-hand movement. For another subject, the imagination of the right-hand movement was replaced by the 'relax' state. For each class, 40 trials were recorded, the class cues being presented on the screen as letters 'L' for left and 'R' for right (respectively 'X' for relax), in a random order. Immediately after the training session and using the classifier computed from it, an online feedback session was carried out. The classification methodology used was band-pass filtering, 8 to 14 Hz, CSP with four output features and as a final stage a linear classifier.

After recording the training data, an 8-fold leave-one-segment-out cross-validation was used to evaluate the level of accuracy achieved in the training data. In the following online feedback session—after possibly adjusting the bias of the classifier—the percentage of trials, where the subject was able to steer the cursor correctly in the indicated direction, defines the accuracy of the feedback phase.

Table 2. Comfort survey.

Subject/age of electrode (months)	Feels better than other dry electrodes tested? (which ones)	Other comments
AA/1	Yes (pins)	
AA/10	Yes (pins)	The pressure of the electrodes and the mounting frame found annoying after 1 h of use
AA/1	Yes (pins)	New test after recoating: after adjusting the pressure for good signal, the pressure is still unobtrusive
AB/1	– (No other sensors tested before)	Seems okay for the whole day
AC/1	Yes (pins, capacitive)	
AC/10	Yes (pins, capacitive)	The pressure of the electrodes found annoying after 1 h of use (as far as the sensor is concerned, these ones are the least bad ones)
AC/1	Yes (pins, capacitive)	New test after recoating: after adjusting the pressure for good signal, the pressure is still unobtrusive
AD/7	Yes, about the same with the capacitive ones (pins, capacitive)	Annoying after some time, not in all versions
AE/10	Subject undecided (capacitive)	Prickling sensation, 'no sensor is good yet!'. No experiments performed
AF/1, AG/1	Irrelevant	Not enough pressure could be applied, the sensation was good but no good contact was achieved
AH/1	No (pins for just 3 minutes)	The subject prefers the pin-based electrodes, mentions prickling sensation

3. Results

The main results are summarized in table 1. The occipital alpha classification produced cross-validation accuracy results between 79% and 92%. All N100 effects were statistically significant (t-test p-value $< 10^{-2}$). All P300 effects were also statistically significant (two-sample t-test p-value $\leqslant 10^{-2}$). The BCI sessions were all successful, just that whereas the subject AD was known to be BCI proficient, the subjects AA and AC were known not to be; therefore, the BCI session for these two subjects were carried out with real executed movements.

4. Discussion

Whereas the signal recorded with the electrode is close to that recorded with a gel-based one only in a certain frequency band (conservatively 7 to 44 Hz, although in many experiments this range was wider), it was enough to warrant the success of the ERP and ERD experiments. As far as the motor imagination BCI is concerned, this frequency range is very good. For ERP-based BCI systems (including the popular P300 implementation), a good behavior from 2–3 Hz would be desirable.

The divergence at low frequencies between the signals provided by our new dry electrode and the gel-based electrode corresponds to the dc drift that affects other new electrodes as well [13].

To analyse to what extent the performance in BCI matches the performance with the gel-based electrodes, we analysed the collection of 25 gel-based training sessions previously recorded with the subject AD. We restricted those recordings to the same subset of channels that we also used in our experiments with the bristle-sensors. We projected each recording onto the reduced set of electrodes and computed the cross-validation training accuracy. The average resulted in 0.78 with a standard deviation of 0.10 . The BCI performance of the same subject in the experiment with bristle-sensor electrodes ($78.8 \pm 13.6\%$) is fairly central to this distribution.

4.1. Comfort survey

As the subjects were members of our labs, most had previous experience of traditional gel-based electrodes and other dry electrodes, such as pin arrays [1, 17, 18] and capacitive ones [11]. We asked the subjects whether or not—in their opinion—the bristle-sensors electrodes are better than the previous electrodes they had experienced previously and we collected their other comments regarding the comfort issue. The collected answers, opinions and direct quotes are given in table 2.

We have performed no experiment involving wearing bristle-sensors longer than 1 h.

Most subjects agreed that the new electrodes are an advance over the pin-based dry electrodes. They suggested ways to improve the bristle-sensor, such as testing geometrical patterns that have a chance of aligning better to the scalp and using even more flexible bristles, as soft as possible (the coating, while flexible, stiffened the bristles to some extent). It is interesting to note that recoating after 10 months of use helped to reduce the pressure that needs to be applied for a good signal (as subjectively felt by the two subjects who were tested just before and just after recoating).

Other ways to improve them would be to add impedance transducers to them and use them thus as active electrodes. For active electrodes, contact with the scalp is less critical to the resulting impedance; therefore, the pressure that currently still needs to be applied could be lowered even further.

5. Conclusions

In this contribution, the results of using new, coated polymer bristle electrodes in the acquisition of the EEG in paradigms related to BCI have been reported. The bristle-sensors produce a signal that compares well with that recorded with gel-based electrodes between at least 7 and 44 Hz. They furthermore provide high redundancy acting towards maintaining mechanical and electrical contact, while providing better comfort to some of the subjects than both the gel-based electrodes and the array-of-pins electrodes, creating thus the opportunity for longer term use in BCI applications.

Acknowledgments

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