33rd Annual International Conference of the IEEE EMBS
Boston, Massachusetts USA, August 30 - September 3, 2011

Performance of Dry electrode with Bristle in Recording EEG rhythms across Brain State Changes

Vangu Kitoko, Tuan N. Nguyen, Jordan S. Nguyen, Yvonne Tran and Hung T. Nguyen, *Senior Member, IEEE*

Abstract—**In this paper we evaluate the physiological performance of a silver-silver chloride dry electrode with bristle (B-Electrode) in recording EEG data. For this purpose, we compare the performance of the bristle electrode in recording EEG data with the standard wet gold-plated cup electrode (G-Electrode) using two different brain state change tasks including resting condition with eyes-closed and performing mathematical task with eyes-open. Using a 2 channel recording device, eyes-closed command data were collected from each of 6 participants for a period of 20sec and the same procedure was applied for the mathematical calculation task. These data were used for statistical and classification analyse. Although, B-electrode has shown a slightly higher performance compared with G-electrode in both tasks, but analyse did not reveal any significant differences between both electrodes in all six subjects tested.**

I. Introduction

An important monitored bio-potential signal is the electroencephalogram (EEG) waves, which refers to the non-invasive measurement of electrical activity produced by the brain as recorded from electrodes placed on the scalp. The physiological explanation of EEG oscillations can be described as the presence of activity in the ascending reticular activating system to stimulate a large number of neurons in the cerebral cortex via the thalamic system, which leads to producing cortical rhythmic activity in very low amplitude range (typically 1-100 μV) and very low frequency behaviour ranged into four band (delta (1-3Hz), theta (3.5-7.5 Hz), alpha (8-13 Hz) and beta (13.5-40 Hz)) depending, on the degree of activity in respective parts of the cerebral cortex, and to the brain state changes (wakefulness, sleep and coma) [1]. Much of the time, these oscillations are irregular, and no specific pattern can be discerned in the EEG and at other times, distinct patterns do appear; however, by application of sensory stimulation, the generators producing EEG activity are coupled and act together in coherent way or periodic [2]. Since alpha waves were traditionally considered as a thalamo-cortical rhythm, it was assumed that cortical arousal was linked to the power and frequency of the alpha rhythm, with high amplitude low frequency activity associated with low cortical arousal and low amplitude high frequency activity associated with high cortical arousal [3], [4]. Additionally, previous studies have reported that peak alpha power at 10 Hz oscillations are generated with great variability every 5–20 sec and last for 0.5–10 sec [5].

The authors are with the Key University Research Centre for Health Technologies (CHT), Faculty of Engineering, University of Technology, NSW 2000, Australia (e-mail: Vangu.Kitoko@student.uts.edu.au; Phone: +61 2 9514 7531)

Despite these promising results, the usability of upper power alpha as cognitive behaviour for longer periods of time EEG recording does not necessarily imply underlying oscillatory activity at that frequency. This is because non-oscillatory, large-amplitude artefacts and transient signals can also produce power increases at specific frequencies [6]. Therefore, some factors should be considered in new designs and during long-term recording sessions to acquire good EEG signals with safety, higher signal to noise ratio (SNR) and no data loss. In accord with Usakli [7], choosing the correct electrode and successful electronic design strategy are the important issues to consider. However, given the recent advances in miniaturised and wireless EEG monitoring technologies, progress in the electronic issue has been already undertaken [8], [9], but the successful to choosing the correct electrode for long-term recordings of scalp potentials is hardly investigated yet.

In general two major types of scalp electrodes are used routinely: wet electrodes and dry electrodes. When using wet electrodes, if high quality EEG recordings are to be obtained without appreciable delay after the application of the electrodes, it is still common practice to carry out preliminary skin preparation and use of a conductive gel or pasta at the relevant locations to reduce the natural skin resistance. This is time consuming and inconvenient [10] and effort to overcome these disadvantageous has led to dry electrodes [11], [12]. As dry electrodes do not require gel or paste, the skin-electrode electrical contact is created by exerting mechanical pressure on the electrode to continue being in tight contact with the scalp. However, this procedure is counterbalanced by the difficulty in making electrode firm mechanical and electrical contact to the scalp due to the presence of hair and subject comfortability. Generally, the electrode fall off and, hence, require the shaving of the scalp site prior to electrode application, which is unpleasant for the subject.

With these issues in mind, two recently works, [13] and [14], have therefore introduced the disposable EEG cup electrodes with bristles (see figure 1). Both works have shown that this electrode is capable of detecting and recording adequate brainwave activities with quality from subjects with haired scalp, without use of conductive pasta or gel and with minimal contact pressure to ensure a good electrical contact with the scalp and subject comfort. However, they conclude that the physiological usefulness of the recorded data using these electrodes is still questionable. It is therefore an object of this paper to present the physiological performance of this electrode in recording EEG data. For this end, we compare the performance of the bristle electrode in recording EEG data with the standard wet gold-plated cup electrode using two different thought pattern tasks including relaxing with eyes-closed and performing mathematical task with eyes-open. The collected data for both electrodes in comparison were then used for statistical and classification analyses.

II. MATERIALS AND METHODS

A. Electrode with Bristles

This electrode is a reusable silver/silver chloride made with twelve 2 *mm* contact posts or bristles for EEG testing without need for scalp preparation and use of conductive gel or pasta (Figure 1A). To become functional, this disposable requires a cup retaining clip lead (Figure 1B), which ensures a good electrical contact and provide high quality brainwaves by improving the attachment of lead-electrode interface.

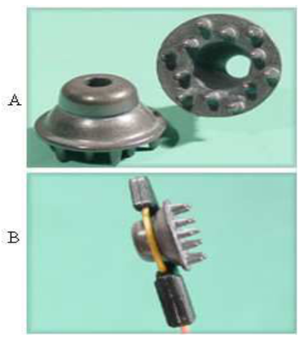

Fig. 1. (A) Electrode with bristles and (B) Bristle electrode with clip lead.

Constructionally, the presence of multiple small pointed projections (or bristles) are distributed over the surface area of electrode so that when the electrode is appropriately applied to the scalp skin and held firmly in contacts using headband or headgear, the bristles will penetrate thick hair and press into the skin without penetrating beyond the epidermis. Although there is no penetration of the stratum corneum, but the presence of the edges of the bristles in firm contact with skin causes the skin to be stretched around the edges so that the stratum corneum at these edges is appreciably thinned, providing a good electrical contact between the electrode and the scalp without appreciable delay after the application of the electrodes and without need for any preliminary scalp-skin preparation and use of conductive gel or pasta. Importantly, it is possible to ensure that only small contact pressure was needed to obtain a robust mechanical fixation of the electrode on the scalp skin and the same time avoid penetrating. As result, the longer term high quality brainwaves detection was performed in a most convenient way, without subject cutting the hair and discomfort (Tactile sensations or pain).

B. Participants

We collected EEG data from eight (8) healthy subjects (6 males including 2 dark-skinned and 4 fair-skinned, and 2 females including 1 dark-skinned and 1 fair-skinned). This is because we have noted that EEG signals can tend to be difficult to generalise from person to person, depending to the skin impedance and working environmental conditions in use [15]. What may work well on a subject with low skin impedance in a warm and humid environment may be found later to fail on a high impedance subject, especially in a cold or dry environment.

All participants were volunteers and had a similar educational background, taking no medication, and reporting no medical treatments or health problems. The experiment was undertaken at the Centre for Health Technologies (CHT) with the understanding and written consent of each participant, following the recommendations of the ethics committee of the University of Technology, Sydney-Australia.

C. Electrode Placement

In all eight participants, the EEG activity was monitored using a 2-channels wireless system developed at the CHT to acquire wireless EEG signals in real-time [8].

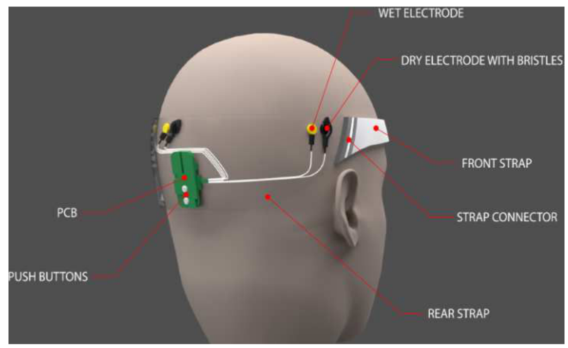

Fig. 2. Illustration of electrode placement for the 2-channels used.

Hence, selected 2-channel EEG montages shown in figure 2 were applied on its ability to detect brainwaves identified on 10/20 electrode placement system as follows: the bristle-electrode for EEG data recording, named B-active electrode was placed over the left occipital region (O1) because of a strong biological signal (alpha) could be clearly identified; its reference electrode also was a Bristle-electrode (B-Reference) placed at T4. The combination of both represents the B-channel. Similarly, the standard gold cup EEG electrode used as sensor is named G-active electrode (9 mm of external and 7 mm of internal diameters) and was placed 5 mm lateral to the position of the B-active. Its reference was also a standard electrode (G-Reference) and placed at 5 mm to B-Reference, which is combined with G-active to make the G-channel. The close position between the G-electrodes and B-electrodes is justified by the fact that it was adequate for the qualitative comparisons of the EEG signals obtained between both types of electrodes. For all eight subjects, we used the G-electrodes with gel or pasta; while the B-electrodes were used without gel and the four electrodes used were held in contact with scalp using a headband and their impedances were maintained very low (5-10KΩ) [16].

D. Procedure and Data collection

EEG data collection was based on two different thought tasks including the brain activity test of awake and resting person, with the eyes closed and the mind wandering, and person performing mathematical calculations. In each session, subject was told that they would be participating in an EEG study and was therefore asked to completely relax in an office chair with eyes closed whereas the EEG was recorded for a period of 20sec. The amplified signal is digitised via an analogue-to-digital converter, with sampling rate of 256 Hz. The digital EEG signals are then low-pass filtered (0.1Hz) to remove the spike component contaminated by high-frequency artefacts due to respiration and cardiac variations [13] and pass the lower frequency signals to be displayed as absolute spectral power into four frequency bands for analysis, delta (1-3Hz), theta (3.5-7.5 Hz), alpha (8-13 Hz) and beta (13.5-40 Hz).

All subjects were initially tested, however two (2) male subjects (1 dark-skinned and 1 fair-skinned) were excluded from the study due to particularly thick resulting to no continuously stable signal could be extracted. As per our standard EEG procedure involving skin preparation and use of conductive pasta and also use of electrode with bristle, subjects were instructed to end the session if they felt any discomfort (minor scratch, irritation, tactile sensations or pain). No injury of any kind occurred and no serious discomfort was reported.

III. RESULTS

As a first result, power spectra were presented for B-electrodes in comparison with G-electrodes. Both electrode recordings were obtained at O1 while the subject-1 is at resting condition with eyes closed for around 20s. We detected prominent peaks in the alpha frequency band around 10 Hz and also in the beta range (15-25Hz), around 20 Hz for both (B and G) channels (Figure 3).

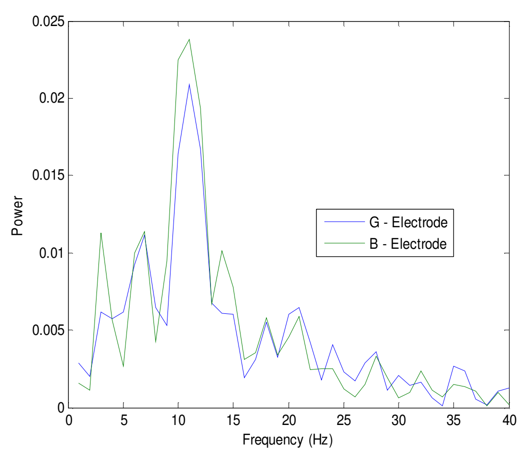

Fig. 3. A comparison of EEG activity recorded with B-electrode and G-electrode for eyes-close task.

For confirmation, the same event was experimentally repeated also to the remaining five subjects. Their spectral profiles, not included here, showed optimal match between both G-electrode and B-electrode in all tested subjects similar to figure 3.

To quantify the comparison structure of the spontaneous alpha oscillations between both channels, we used a power spectrum analysis, which measures the contribution of different frequencies to the total power (P) of a signal. Using the spectrum window theory and the Parseval's theorem [17], the total power in the alpha band is considered to be the sum squared of power distribution values at a discrete frequency and given as:

$$P = \sum_{i=0}^{N-1} |p_i|^2 \qquad (1)$$

where N is the length of window function (p) for the power spectra estimation in the signal used and defined by a vector of real numbers $\{p_i\}$. Because we could assume that all six alpha power spectra windows studied were symmetric, only the $N/2+1$ coefficients $p_0, \dots\dots, p_{N/2}$ were needed to compute the total power in alpha range across subjects for B and G electrodes using the equation 1. Results obtained are presented in Table 1, which have shown that the average values over all six subjects tested (Mean: 0.000877 for G-electrode and 0.001112 for B-electrode) did not reveal any significant differences between both electrodes in all six subjects tested.

TABLE I

TOTAL POWER FOR ALPHA BAND (8-13Hz) OVER ALL SIX SUBJECTS

Participant	G - Electrode	B - Electrode
Subject-1	0.000791	0.001109
Subject-2	0.001022	0.001342
Subject-3	0.000705	0.000922
Subject-4	0.000874	0.001139
Subject-5	0.000890	0.001017
Subject-6	0.000982	0.001145
Mean	**0.000877**	**0.001112**

The power spectrum analyses used in the previous section are not optimal for the quantification of B-electrode performance in comparison with G-electrode. For optimal robustness of the evaluation, the session with Brain-computer interface (BCI) was performed with B and G electrode using two mental tasks: eye-closed and mathematical calculation, as presented in figure 4. It aimed at showing the possibility of performing BCI experiment with the B-electrodes and not at comparing them with G-electrodes. For this end, the combined EEG raw data were first transformed in FFT domain and the EEG powers at frequencies from 1 to 40Hz were then fed into Multilayer Feed-Forward Neural Network. By trial and error, 25 hidden nodes were chosen, and only one output node was required for two-task classification (0 for arithmetic calculation task and 1 for eye close task). For two-mental task classification (math calculation and eye-close), the overall accuracy of the Neural Network trained with EEG data collecting from B-electrodes was 99.79%, whereas the accuracy of the Neural

Network trained with EEG data collecting by G-electrode was 98.084%.

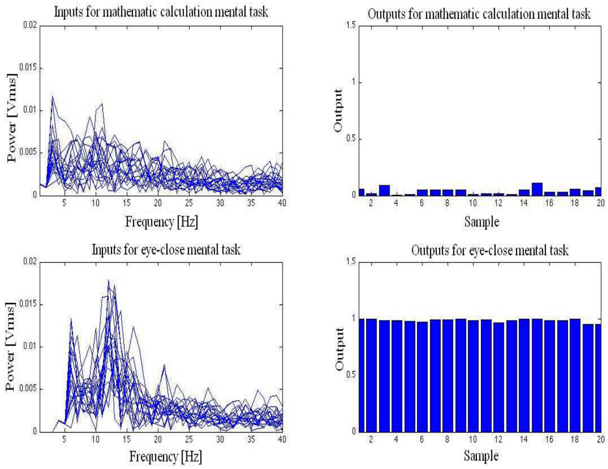

Fig. 4: Eye-closed and mathematical calculation mental tasks using B-electrode.

IV. CONCLUSION

In this research work we have evaluate the physiological performance of the B-electrode in recording EEG data compared with the standard wet G-electrode. For robust optimal evaluation, two different thought pattern tasks including closing eyes and performing mathematical calculation have been used. We conclude this paper by pointing out the following:

1. The B-electrode is capable of recording EEG with a signal quality similar to that obtained with the wet G-electrode without need for any preliminary scalp-skin preparation and use of conductive gel or pasta and with only small scalp-electrode contact pressure to obtain a robust mechanical fixation and the same time avoid penetrating. Importantly, the high quality brainwaves detection was performed in a most convenient way, without subject cutting the hair and discomfort.

2. Although the performance of B-electrode slightly higher than G-electrode in all six subjects tested with closing eyes experiment, we point out that both, statistical and classification, analyses did not reveal any significant differences for EEG power spectral responses data recorded.

3. Although results demonstrate a number of advantages of the B-electrode compared to G-electrode, we are continuing to develop advanced technology to improve the accuracy of signal for long-term and ambulatory EEG recording.

ACKNOWLEDGEMENT

Authors thank Greg Hunter (Centre for Health Technologies-UTS), Raymond Clout (Research Laboratories -UTS) and Nejhdeh Ghevondian (AIMEDICS Pty Ltd) for their supports.

REFERENCES

[1] H. Nguyen, Electroencephalogram, “Biomedical Instrumentation and Applications,” University of Technology, Sydney – Australia, 2008.

[2] T.A. Whitten, A.M. Hughes, C.T. Dickson and J. B. Caplan, “A better oscillation detection method robustly extracts EEG rhythms across brain state changes: The human alpha rhythm as a test case,” *NeuroImage,* vol 54, pp. 860-874, 2011.

[3] S. Hanslmayr, W. Klimesch, P. Sauseng, W.R. Gruber, M. Doppelmayr, R. Freunberger and T. Pecherstorfer, “Visual discrimination performance is related to decreased alpha amplitude but increased phase locking,” Neuroscience letters, vol. 375, pp. 64-68, 2005.

[4] Y.Tran, A. Craig, P. Boord, K. Connell, N. Cooper and E. Gordon, “Personality traits and its association with resting regional brain activity,” Internationational journal of psychophysiology, vol. 60, pp. 215-224, 2006.

[5] K. Linkenkaer-Hansen, V.V. Nikouline, J.M. Palva and R.J. IImoniemi, “Long-range temporal correlations and scaling behaviour in human brain oscillations, Neuroscience, vol. 21, pp. 1370-1377, 2001.

[6] B. Zoefel, R.J. Huster and C.S. Herrmann, “Neurofeedback training of the upper alpha frequency band in EEG improves cognitive performance,” Neuroimage, vol. 54, pp. 1427-1431, 2011.

[7] A. B. Usakli, Improvement of EEG signal acquisition: An electrical aspect for state of the art of front end. Hindawi Publishing Corporation - Computational Intelligence and Neuroscience, 7 pages, 2010.

[8] H.T. Nguyen and T.W. Jones, “Detection of nocturnal hypoglycemic episodes using EEG signals,” 32nd Annual International Conference of the IEEE EMBS-Buenos Aires, Argentina, 2010.

[9] A.J. Casson, D.C. Yates, S.J.M. Smith, J.S. Duncan and E. Rodriguez-villegas, “Wearable electroencephalography,” IEEE Engineering in Medicine and Biology Magazine, Vol. 29, pp. 44-56, 2010.

[10] P. Griss, P. Enoksson, H.K. Tolvanen-laakso, P. Merilainen, S. Ollmar and G. Stemme, “Micromachined electrodes for biopotential measurements,” Journal of microelectromechanical systems, vol. 10, pp. 10-16, 2001.

[11] G. Ruffinia, S. Dunnea, L. Fuentemillab, C. Graub, E. Farres, J. Marco-pallare, P.C.P.Watts & S.R.P Silvac, “First human trials of a dry electrophysiology sensor using a carbon nanotube array interface,” sensors and Actuators A, vol. 144, pp. 275-279, 2008.

[12] C. Fonseca, J.P. Silva cunha, R.E. Martins, V.M. Ferreira, J.P. Marquesdesá, M.A. Barbosa, and R.E. Martins. A Novel dry active electrode for EEG recording. IEEE Trans. on Biomed. Eng., vol. 54, pp. 162-165, 2007.

[13] A. Chatila, J.V. Asdlan, J. Bitz, H. Barrow, M. Fimbres and M. Crowder. Portable brainwave monitor. Interdisciplinary engineering design program, The University of Arizona – USA, 2009.

[14] L. Brown, J.V.D. Molengraft, R.F. Yazicioglu, T. Torfs, J. PENDERS and C.V. Hoof, "A low-power, wireless, 8-channel EEG monitoring headset," 32nd Annual International Conference of the IEEE EMBS, pp. 4197-4200, 2010.

[15] E. McAdams, "Bioelectrodes," in Encyclopedia of Medical Devices and Instrumentation, Second Edition, J. G. Webster, Ed., ed: John Wiley &Sons, Inc., pp. 120-166, 2006.

[16] Y. M. Chi, T.P. Jung and G. Cauwenberghs, “ Dry-contact and noncontact biopotential electrodes: Methodological review. IEEE Reviews in biomedical engineering,, vol. 3, pp. 106-119, 2010.

[17] G. Heinzel, A. Rüdiger and R. Schilling, “Spectrum and spectral density estimation by discrete fourier transform (DFT), including a comprehensive list of window functions and some new flat-top windows,” Albert-Einstein-Institut, Hannover, 2002.