EEG Acquisition System based on Active Electrodes with Common-Mode Interference Suppression by Driving Right Leg Circuit

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***Abstract*— We present a system for the acquisition of EEG signals based on active electrodes and implementing a Driving Right Leg circuit (DgRL). DgRL allows for single-ended amplification and analog-to-digital conversion, still guaranteeing a common mode rejection in excess of 110 dB. This allows the system to acquire high-quality EEG signals essentially removing network interference for both wet and dry-contact electrodes. The front-end amplification stage is integrated on the electrode, minimizing the system's sensitivity to electrode contact quality, cable movement and common mode interference. The A/D conversion stage can be either integrated in the remote back-end or placed on the head as well, allowing for an all-digital communication to the back-end. Noise integrated in the band from 0.5 to 100 Hz is comprised between 0.62 and 1.3 μV, depending on the configuration. Current consumption for the amplification and A/D conversion of one channel is 390 μA. Thanks to its low noise, the high level of interference suppression and its quick setup capabilities, the system is particularly suitable for use outside clinical environments, such as in home care, brain-computer interfaces or consumer-oriented applications.**

I. INTRODUCTION

Electroencephalogram (EEG) recording offers the capability to collect information on the activity of the brain with some significant advantages over other techniques (such as fMRI, PET, MEG, ECoG), which can be summarized as being the least invasive, the easiest to setup and the one with the lowest cost [1][2][3][4]. In recent times, the fields of EEG application have widened from those typical of the clinical setting to include Brain-Computer Interfaces (BCI) and also consumer-oriented applications ranging from home care to neurofeedback and gaming controllers. There is therefore a demand for high-quality acquisition systems whose montage is fast and does not require trained personnel. Systems based on dry electrodes are particularly attractive since they make for very simple, almost instantaneous setup [1].

One of the main issues affecting the quality of the signals extracted by an EEG acquisition system (Fig. 1(a)) is interference (such as from power-lines) which can either generate displacement currents in the leads from the electrodes to the acquisition system or generate common mode voltages on the subject which are converted to differential signals if the Common Mode Rejection (CMR) of the system is insufficient. Active electrodes systems (Fig. 1(b)) [2][3][4] integrate an amplification stage directly on the electrode with the purpose of providing the electrode with a very high load impedance while driving the cable with a low impedance, making the system more robust to interferences on the leads and to the potential divider effect at the input. In order to improve the system's ease of use, minimization of the number and size of cables is fundamental. One possible approach may be to move the analog-to-digital conversion stage close to the electrode, so as to remove any need for noise-sensitive analog signals being routed from the electrodes to the back-end (Fig. 1(c)) [4]. For the same

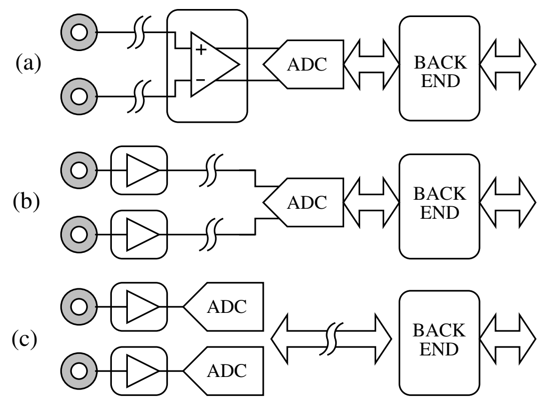

Fig. 1. (a) Standard EEG acquisition system, (b) EEG system with active electrodes locally amplifying the signal, (c) EEG system with local amplification and A/D conversion of the signal.

reason, single-ended acquisition is usually preferred in active electrode systems, since differential acquisition calls for additional wires to be routed to every electrode. This leads to requirements on the gain accuracy of the amplification stages and, in some cases, the gain itself needs to be limited to a few units, increasing the noise contribution of the following stages [2]. For these reasons, CMR performance is often insufficient and additional circuitry is required in order to reduce the common mode interference on the patient and thus improve CMR performance of the system. This ranges from standard Driven Right Leg (DRL) circuits (which suffer from stability issues [5]) to complex Common Mode Feedback (CMFB) techniques (requiring additional wires to be routed [3][4]) which do however provide a limited improvement.

II. SYSTEM DESCRIPTION

We present a flexible implementation of the system in Fig. 1(b)(c) which makes use of a recently introduced Driving Right Leg (DgRL) [6] scheme (see Fig. 2-left) where the common mode interference V_{CM} is detected by the reference electrode (REF) and directly drives a potential V_R, fixed with respect to the isolated ground of the instrumentation

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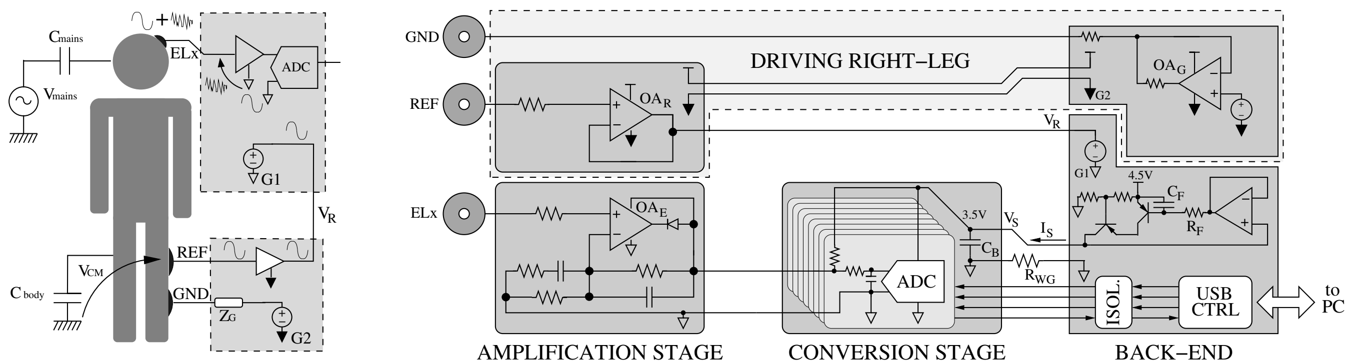

Fig. 2. On the left: Simplified representation of a DgRL-based acquisition system. V_{mains} couples to the subject through C_{mains} and C_{body}, resulting in a common mode interference V_{CM} which is detected by electrode REF. On the right: Simplified representation of the developed system. The overall distance between electrodes and back-end is approximately 1.2 meters.

(G1). V_{G1} therefore follows the common mode intereference. In this way, a single-ended stage will inherently act as differential and will encounter very little common mode at its input, automatically rejecting most of the interference and loosening requirements on the maximum value and accuracy of the amplification stage's gain. Moreover, since the input impedance of the amplifier will be directed towards the reference signal, potential divider effects at input will be significantly reduced. The system described is the first implementation of a DgRL circuit to active electrode systems and it shows how the levels of CMR which can be obtained are high enough to allow the system presented to acquire high-quality EEG signals essentially removing network interference without the need for skin preparation for both wet and dry-contact electrodes.

The acquisition system is depicted in Fig. 2-right. Depending on the configuration (as in Fig. 1(b) or (c)), the length of the cables connecting the three sections will vary. In all cases, A/D conversion is performed for groups of 8 electrodes by 8-channel ADCs. Depending on architecture choices, a single-channel ADC could be placed directly on the electrode, obtaining benefits in terms of system flexibility, especially in the case of failure by a single electrode. In this work, we chose to always adopt the same 8-channel ADC for the sake of coherence and, more importantly, to minimize the number of cables between the subject's head and the back-end, simplifying connections and making the system more compact, lighter and easier to handle.

A. EEG signal acquisition

Signal amplification is performed on the electrode by a low-power, low-noise, rail-to-rail Operational Amplifier (O.A.) connected in non-inverting configuration (OA_E in Fig. 2). Protection resistors are used in order to limit patient auxiliary current in case of system faults. The O.A. is an OPA378 from Texas Instruments, which has a quiescent current of 125 μA and low current and voltage noise, with no significant increase at low frequency. In order to minimize the number of wires, the output signal of the amplification stage is used for providing power supply to the O.A. as well. A forward-biased diode connects the output pin of the O.A. to the power supply. The output is biased by the following stage which entails a 6.8 kΩ resistor toward a 3.5 V power supply. The output of the amplification stage can swing between 2 V, which is the minimum power supply voltage of the O.A. and 2.4 V which is the maximum input voltage of the A/D conversion stage. The amplification stage is configured for a DC gain of 1.12 and an AC gain in the EEG signal bandwidth (0.5 to 100 Hz) of approximately 45, which is enough to reduce the input referred noise of the 16-bit A/D conversion stage to acceptable levels. Considering a maximum input signal of $\pm$100 μV, the input DC voltage range is 1.96 V $\pm$ 174 mV, which is enough to accommodate polarization and offset voltages of gold and Ag/AgCl electrodes [1]. The PCB size is 10 mm x 10 mm, compatible with integration on high-density EEG systems.

The Analog/Digital conversion is based on a low-power, 8-channel, 16-bit analog front-end for biopotential measurements (ADS1198 from Texas Instruments). Low weight ($\approx$2 gr/m) 1-mm diameter coaxial cables connect the active electrode to the ADC-section, with the shield carrying ground connection and the core for both power supply and active electrode output. The electrode signals are low-pass filtered by a passive RC-filter with a 16 kHz bandwidth and sent to the non-inverting inputs of the ADC. In order to avoid the need to generate an additional reference signal (and add its noise contribution to that of the acquisition chain), inverting inputs are tied to ground and the input range of the ADC is comprised between 0 and 2.4 V. The back-end exchanges data between the ADCs and the PC and provides power supply to the ADC boards. A standard 6-core ribbon-cable (variable length between 2 and 120 cm) connects the ADC board to the back-end, carrying power supply and ground, plus the four digital signals for SPI data transfer. In the implementation of Fig. 1(c), due to significant return currents on the ground connection (especially but not confined to supply current to the digital interface) and resistance R_{WG} $\approx$1 Ω associated with the 1.2 m cable, the voltage drop on R_{WG} can easily increase the noise of the system to unacceptable levels. In order to avoid this, bypass capacitors C_B to an overall value of approximately 100 μF are placed between power supply and ground, and the circuit depicted in Fig. 2-right is used for providing power supply. Its objective is to act as a voltage source with an output resistance having a low value at DC (in order to minimize V_S variation as a function of fluctuations in the supply current I_S at DC) and a significantly higher value from the EEG band on. At very low frequency (below the frequencies of typical EEG band), the circuit acts as a voltage source, with an output voltage of $V_S = 4.5 - V_{EB,on} - (R_F/\beta_0) \cdot I_S \approx 3.5$ V and an output resistance of approximately $R_F/\beta_0 \approx 150\ \Omega$, where $V_{EB,on} \approx 0.7$ V is the emitter-base voltage of the BJT and $\beta_0 \approx 100$ its DC current gain. At higher frequencies ($>$ 0.1 Hz), the RC filter reduces the open loop gain of the circuit, gradually increasing the output impedance up to that of a cascode current source (> 100 kΩ) so that most of the return currents on the ADC will close on the on-board bypass capacitors rather than on the back-end.

B. Driving Right Leg Circuit

The Driving Right Leg circuit has a separate isolated ground G2 and power supply from the rest of the system and is based on two electrodes (REF and GND). REF is the active-electrode with unitary gain which detects the reference voltage and is connected to a voltage reference which keeps the potential difference ($V_R - V_{G1}$) to a fixed value, forcing V_{G1} and all fixed voltages to follow the potential on the REF electrode. It should be pointed out that, since this is the only low-impedance connection between the acquisition section and the DgRL, only a very small current will flow across the connection itself. The ground electrode GND fixes the potential of the subject with respect to the isolated ground G2 in order to avoid saturation of OA_R. Protection resistors are used in order to limit patient auxiliary current in case of system faults. Due to the finite output impedance, the potential on the subject will contain residual interference from the mains (with respect to isolated ground G2). Both OA_R and OA_G are OPA378 from Texas Instruments.

III. MEASUREMENTS

A. Electrical Characterization

Noise performance is tested by shorting the inputs of the active electrodes. When the ADC is placed close to the back-end, the system shows a noise level of 0.98 μV$_{RMS}$, integrated in the band between 0.5 and 100 Hz, with no residual 50 Hz interference. If one computes the difference between two channels in the digital domain after analog-to-digital conversion, the integrated noise decreases to 0.61 μV$_{RMS}$, due to the removal of common mode noise components. When the ADC is moved closer to the electrodes, the system shows an increase of noise level up to 1.3 μV$_{RMS}$, due to some residual digital noise which leads to spikes on the local ground of the ADC board caused by the voltage drop on R_{WG}. These results, though slightly higher than the 0.5 μV$_{RMS}$ recommended by IFCN [7], are compatible with high-quality EEG recording and in line with or lower than previously reported active electrode systems [2][3][4].

In order to evaluate the CMR performances as prescribed by standard IEC-60601-2-27 [8], the circuit in Fig. 3-top

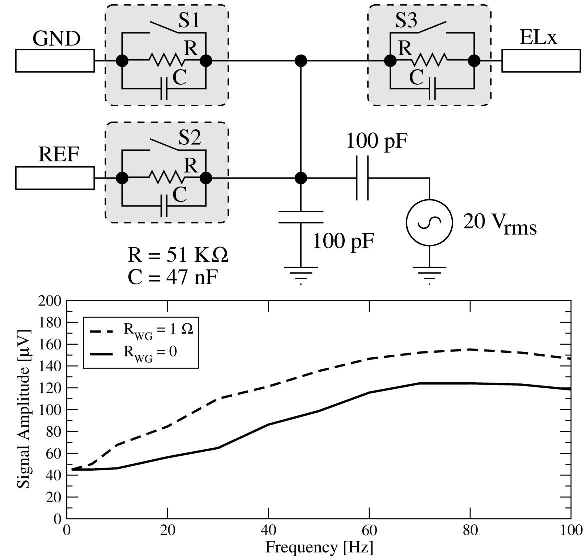

Fig. 3. Test setup (top) and results (bottom) for CMR performances as prescribed by standard IEC-60601-2-27 [8], as a function of frequency.

is connected to the inputs of the EEG acquisition system. Electrodes are modeled using resistor R and capacitor C connected in parallel as illustrated. Three switches (S1 to S3) allow one to short these impedances, so as to simulate a complete mismatch in the signal differential path. The standard prescribes that, for a 50/60 Hz sine wave excitation signal with an amplitude of 20 V_{RMS}, and considering a complete mismatch in the signal differential path (either S2 or S3 closed), the output signal should be below 1 mV peak-to-peak, corresponding to a CMR of 95 dB with respect to the mains signal. Fig. 3-bottom shows CMR performances for the two systems, with the ADC board close to the back-end (Fig. 1(b), $R_{WG} \approx 0$) or to the electrode (Fig. 1(c), $R_{WG} \approx 1\ \Omega$) at frequencies between 1 and 100 Hz. Performances are very similar, with a maximum value of the residual common mode interference of approximately 150 μV, a value which is more than 6 times smaller than the upper limit defined by the standard.

Current consumption of the different sections are:

Amplifier	190 μA per ch.
ADC	200 μA per ch.
Back-end	4 mA
DgRL	$<$ 1 mA

B. Functional Characterization

Fig. 4 shows results for EEG acquisition on electrode O2 with respect to Fp2, which corresponds to the REF electrode of the DgRL circuit. The ground electrode GND is located in Fpz. The system tested has the ADC close to the electrodes, since it has been shown to be the worst-case scenario in terms of noise and CMR. The electrodes are completely unshielded and acquisitions are conducted in a typical electrical engineering laboratory, which can be considered a noisy environment and worse than most home environments. No skin preparation is needed, except a

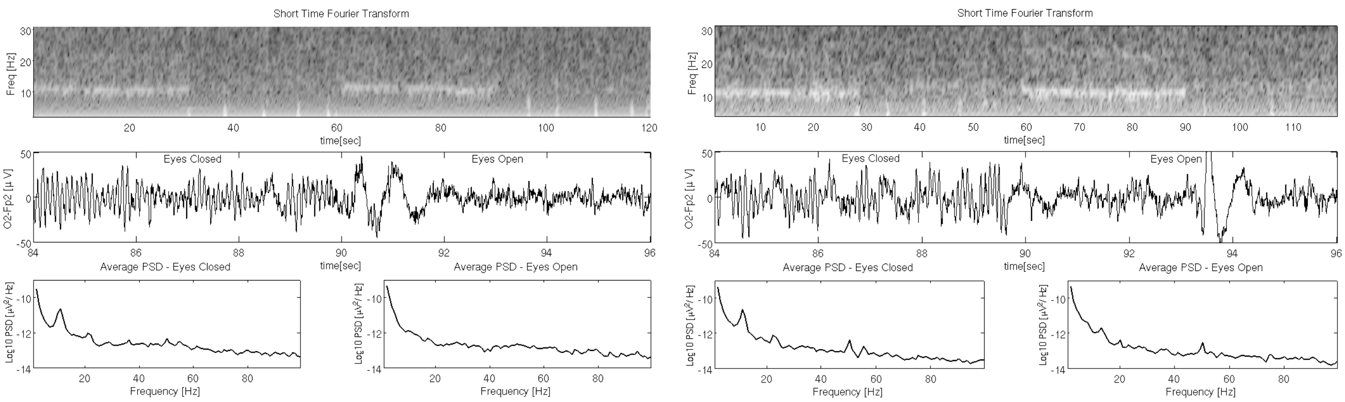

Fig. 4. EEG signal acquisition in O2 with reference in Fp2 for wet (left) and dry-contact (right) electrodes. (a) Short time fourier transform showing spectral content increase in the time intervals where the subject has eyes closed ([0-30] and [60-90] seconds). (b) Signal in the time-domain during eye transition from closed to open, band-pass filtered between 0.5 and 100 Hz, without any filter for 50 Hz suppression. (c) PSD of EEG signal, averaged in Eyes Closed and Open time intervals respectively.

quick and gentle cleaning of the whole head with a dry cloth. For the left part of the figure, the contact with the patient skin is obtained by means of standard 10-mm Ag/AgCl electrodes and conductive paste. On the right, measurements with dry-contact electrodes are presented. These are custom-made electrodes, similar to those in [9], but with only 8 golden-plated spikes and a 12-mm diameter. Electrodes are kept in place by a standard EEG cap. The upper part of the graph shows the Short Time Fourier Transform of the acquired signal which clearly highlights an alpha rhythm at approximately 12 Hz frequency when the subject keeps his eyes closed. This is also visible in the time domain representation of the acquired signal (full EEG band from 0.5 to 100 Hz, without any kind of filtering at 50 Hz), which is presented in the central part of Fig. 4. The eye movement artifacts locate the time instants (time equal to 90 s) at which the subject closes his eyes. After this event, the amplitude of the EEG signal decreases significantly and the alpha rhythmic oscillation disappears. The lower part of the graph shows the Power Spectral Density (PSD) of the signal. The common mode interference from the mains (50 Hz) has been measured to be 2.5 mV_{RMS} for wet electrodes and 5.2 mV_{RMS} for dry electrodes. Thanks to DgRL, the common mode interference at the output of the acquisition system is reduced to less than 1 μV_{RMS} and 1.8 μV_{RMS} respectively (with a reduction in excess of 70 dB) and does not affect the quality of the signal.

IV. CONCLUSIONS

We have presented a system based on active electrodes and a Driving Right Leg circuit improving CMR performances to the point where single-ended amplification and analog-to-digital conversion are possible. The circuit is completely based on low-cost, easily available off-the-shelf components. The system has been tested with both wet and dry-contact electrodes without any skin preparation, proving capable of acquiring high-quality EEG recording, essentially cancelling any common mode interference even in noisy environments. The only disadvantage is the need for a separate isolated power supply which, however, can easily be derived from integrated solutions from the main supply or, given the very limited power consumption, even from small-size batteries (as an example, LR44 batteries could easily provide more than 100 hours of operation). It can be observed how, thanks to its low noise, high level of interference suppression and quick setup capability, the system presented is particularly suitable for use outside clinical environments, such as in home care, BCI or consumer-oriented applications, whilst still providing performances close to its clinical counterparts.

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