High Density Wireless EEG Prototype: Design and Evaluation against Reference Equipment

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***Abstract*— A high density wireless electroencephalographic (EEG) platform has been designed. It is able to record up to 64 EEG channels with electrode to tissue impedance (ETI) monitoring. The analog front-end is based on two kinds of low power ASICs implementing the active electrodes and the amplifier. A power efficient compression algorithm enables the use of continuous wireless transmission of data through Bluetooth for real-time monitoring with an overall power consumption of about 350 mW. EEG acquisitions on five subjects (one healthy subject and four patients suffering from epilepsy) have been recorded in parallel with a reference system commonly used in clinical practice and data of the wireless prototype and reference system have been processed with an automatic tool for seizure detection and localization. The false alarm rates (0.1-0.5 events per hour) are comparable between the two system and wireless prototype also detected the seizure correctly and allowed its localization.**

I. Introduction

The recording and evaluation of the electroencephalogram (EEG) is a well-established technique to evaluate the brain electrical activity and finds application in several fields, from clinical practice to advanced human machine interfaces and entertainment. Some examples include the diagnosis and treatment of epilepsy and sleep disorders, brain computer interface (BCI) and augmented cognition.

Long term EEG monitoring is usually done in hospitals, with an amplifier connected to a recording unit through long cables; video-recording is usually acquired in parallel. Moving to ambulatory or, better, to wearable EEG would improve the comfort for the subject and would enable novel or more pervasive applications of this technology [1].

Wireless EEG is possible, but it usually forces to find compromises in terms of battery lifetime, number of channels, sampling frequency and adopted wireless technology. Data streaming through radio frequency connection (RF), in particular, presents constraints in terms

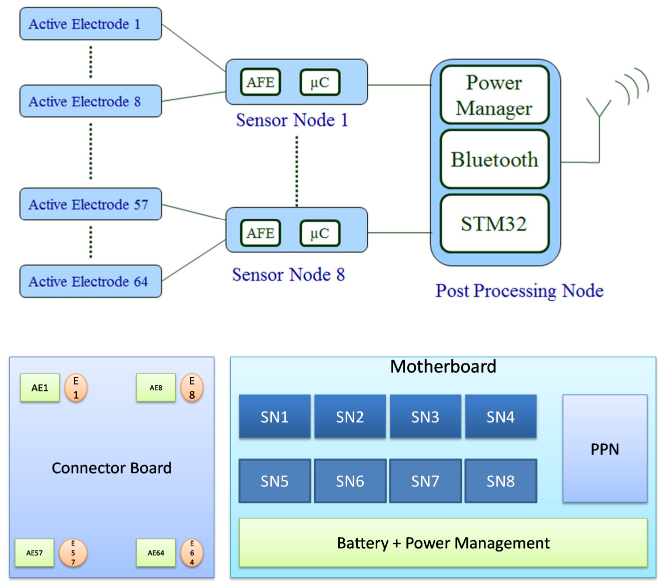

Figure 1. Block diagram of the system (top) and physical implementation

of bandwidth and power consumptions. Devices adopting this approach usually target applications like BCI, in which a reduced number of channels is acceptable [2], or present compromises between sampling rate, number of channels and battery lifetime, like the g.Nautilus device (g.tec medical engineering GmbH) which guarantees 8 hours of recording with up to 32 channels. Memory logging usually enables an extended autonomy like in the TREA device (Grass Technologies). A limit of this approach is that it doesn't allow real-time check of signal quality, forcing the integration of any real-time processing inside the device and preventing any signal quality check. This can become an important limitation in ambulatory recordings, in which an increased amount of movement artifacts or electrodes lead off can be expected. Combined approaches have been proposed for high density and high quality EEG recording for epilepsy monitoring [3]. However RF streaming can be used only for a subset of channels and results in a complementary technology unsuitable for real-time monitoring.

This work aims to demonstrate the feasibility of a high density wireless EEG system (up to 64 channels with at least 600 Hz sampling rate) based on real-time data streaming through RF link. The system is presented in section II, it is based on low power, application specific integrated circuits (ASICs) used for the active electrodes and amplifier stage. It is able to record, together with EEG, electrode to tissue impedance (ETI) and to perform power efficient data compression and transmission through Bluetooth. The first

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prototype has been realized with the aim of demonstrating the general concept and validating the signal quality through first trials in a clinical environment, as described in section III. Results of this first evaluation are discussed in sectionIV.

II. System Overview

A. Hardware description

The design of the wireless system followed a modular approach. The block diagram is showed in Figure 1.

In order to reduce size and power consumption, custom ASICs, described in [4], have been used for the analog front-end. The system consists of two ASICs, active electrode and a readout stage with instrumentation amplifier and integrated 12 bit ADC. Active electrode ASIC is a low power buffer (30 μW) with a very high input impedance. The analog outputs from the active electrodes are connected to the read-out stage. Active electrodes improve the immunity to artifacts and noise coupled on the cables. They also can generate the programmable current needed for the measurement of ETI.

Each group of eight active electrodes is connected to a sensor node (SN), which consists of readout ASIC, an eight channels instrumentation amplifier with integrated 12 bit AD converter and microcontroller (MSP430F1611 from Texas Instruments). The SN provides the signals needed by the connected active electrodes, and, once a start of conversion (SOC) signal is received, it executes data acquisition of the 8 channels, providing a total of 24 samples: 8 EEG, 8 in-phase ETI and 8 quadrature ETI. Eight sensor nodes are integrated on the device, providing up to 64 EEG channels.

The post processing node (PPN) is responsible for system management and synchronization, data compression and wireless transmission. Sampling frequency for EEG and ETI is programmable. In order to design the data rate of the wireless communication, assumptions have been made about maximum data rate in typical scenarios. Considering a sampling frequency of 600 Hz for EEG and of 150 Hz for ETI, 16 bits for the samples and the compression ratio of 2 as described in section B, the data rate ranges from 230.4 kb/s (48 channels EEG only) to 460.8 kb/s (64 channels EEG and ETI). In this scenario, Bluetooth has been judged a very good compromise between data rate and power consumption: its nominal speed reaches 2 Mb/s which provides a sufficient margin over the data rates above. Furthermore Bluetooth is really widespread in many portable devices which can be successfully used as personal device assistant for telemedicine applications. The SPBT2632C2 module (STMicroelectronics) has been selected, it provides a simple interface to microcontroller through string-based commands and reasonable power consumption (between 40 and 80 mA).

A STM32L151 (STMicroelectronics) has been selected as microcontroller; it is based on an ARM Cortex M3 core, with current consumption of 233 uA/MHz. The maximum frequency of 32 MHz and the memory availability (48 kB RAM, 384 kB Flash) are enough to manage data transfer, and to run the compression algorithm described in section B. The PPN provides a SOC signal to all the SNs which triggers a new conversion. At the same time each SN makes the samples of previous conversion cycle available through SPI, so that the PPN can read, in sequence, the data from all the SNs, compress them and send to the host through Bluetooth.

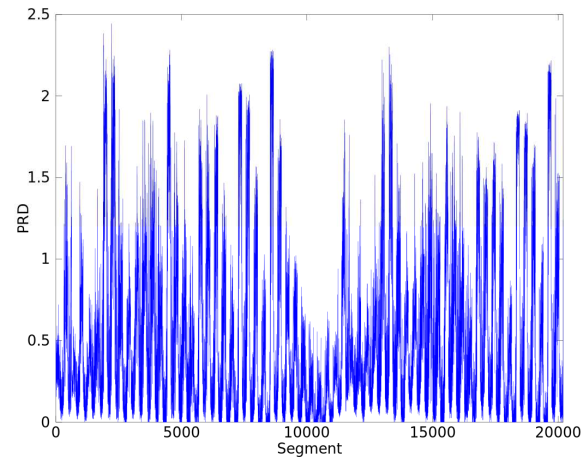

Figure 2. PRD calculated on 3 minutes long segments in a 18 hours 64 channels acquisition.

The system is powered through a 2.9 Ah rechargeable Li-Ion battery (PA-LN19, Panasonic) which guarantees continuous data acquisition for more than 24 hours.

B. Compression algorithm

Reducing the transmitted data rate in a wireless EEG system is crucial in order to reduce power consumption in the RF section, which is usually the higher contribution, and in order to improve the reliability of the wireless connection. The compression algorithm, on the other hand, must not require a too high computational power. Indeed it must be implemented on the microcontroller with reasonable clock speed, in order to avoid increasing too much the power needed for the processing.

Lossless compression is usually preferred for EEG [5], however lossy compression enables a good compression ratio with limited computational power with minimum distortion. Lossy compression techniques commonly used for imaging have been successfully applied to EEG [6]. In this work we used a STMicroelectronics proprietary lossy algorithm for imaging encoding (the same used in [7]) to compress the EEG and ETI signals. The encoder works on a fixed length data buffer and guarantees a fixed compression ratio; this architecture results simple to be managed in the microcontroller firmware, since it enables a complete prediction of the output data amount and of the data rate. A compression ratio of 2 has been selected, since it was judged to be a good compromise between final data rate and distortion level. This was evaluated as percentage root-mean-square difference (PRD), calculated as

$$PRD = \frac{\sqrt{\sum (x_i - \tilde{x}_i)^2}}{\sqrt{\sum (x_i - \bar{x})^2}} \times 100$$

in which x_i and $\tilde{x}_i$ are the samples of the original and of the encoded/decoded signals respectively and $\bar{x}$ is the mean value of the original signal.

The distortion of the algorithm has been evaluated on 18 hours long, 64 channels acquisition. The PRD has been evaluated dividing each channel in segments of 3 minutes. The final result, for all the channels put in sequence, is displayed in Figure 2. The PRD is below 2.5% with a mean value of 0.5%.

III. METHODS

In order to verify the design, the prototype was initially characterized on system level parameters, like gain, bandwidth, input referred noise, etc.

Then, in order to demonstrate the concept and to verify the system in a clinical scenario, the device has been tested on 5 subjects (a healthy subject and 4 patients), in the expertise center for epileptology, sleep medicine and neurocognition of Kempenhaeghe (Kempenhaeghe, Heeze, The Netherlands) in the same environment used in the common clinical activity. The patients were known to have epilepsy and were subject to pre-surgical evaluation via long-term video-EEG monitoring for up to 5 days. Patients who were not able to meet the mild physical or psychological criteria to comply the test procedure were excluded from the study. The study was approved off by the medical ethical committee of Kempenhaeghe. All the subjects were older than 18 and gave a written informed consent

The sessions were performed in the inpatient clinic using the video-EEG acquisition system available at Kempenhaeghe, which was coupled to the newly developed wireless EEG prototype. The patients, during the acquisition, were free to move inside a large living room while EEG signal was recorded in parallel by the two systems. The recordings with the wireless system have been done during the day (6 to 8 hours / day), recharging the device in the night, while the reference system guaranteed continuous monitoring 24 hours / day. Stellate is used as a reference system. The number of electrodes varied from 32 to 48 depending on particular patient conditions.

The system integration and packaging was specifically designed to enable the parallel recordings from the two systems. In particular it was decided to use, in this first stage, standard electrodes and gel instead of dry active electrodes. This approach was adopted in order to avoid any possible interference with standard clinical practices, both in terms of possible artifacts generated by non-standard electrodes on the reference signal and of increased effort for electrode placement. On the other hand active electrodes ASICs were integrated inside the prototype, in order to validate the full chain. Splitter cables were used to connect the electrodes to the wireless and the reference system. The wireless prototype was equipped with standard connectors, each one of them was connected to the active electrode ASIC on the connector PCB. A motherboard PCB hosted the PPN, and all the SNs which were plugged on it (see Figure 1). The connector board was connected on top of the motherboard using cables and they were properly shielded to minimize noise coupling form Bluetooth to analog section. The device was equipped with a button driving an external connector through an optocoupler, in order to enable the connection to the reference system for the synchronization of the two devices before the beginning of the acquisition. A PC was used as receiver, and a serial data logger software with scripting capabilities (Docklight, Flachmann und Heggelbacher) was used to send commands to the device, log the data and monitor the data exchange, performing automatic reconnection in case of temporary out of range events. Logged data were decompressed for signal analysis.

TABLE I. SUMMARY OF THE RESULTS

Subject	Reference system		Wireless Prototype	
	False alarm rate (per h.)	*Seizures (recorded / detected)*	*False alarm rate (per h.)*	*Seizures (recorded / detected)*
Healthy subject	0.431	-	0.27	-
Patient 1	0.388	-	0.108	-
Patient 2	0.383	3/3	0.253	1/1
Patient 3	0.234	1/0	0.486	-
Patient 4	0.190	8/6	0.268	-

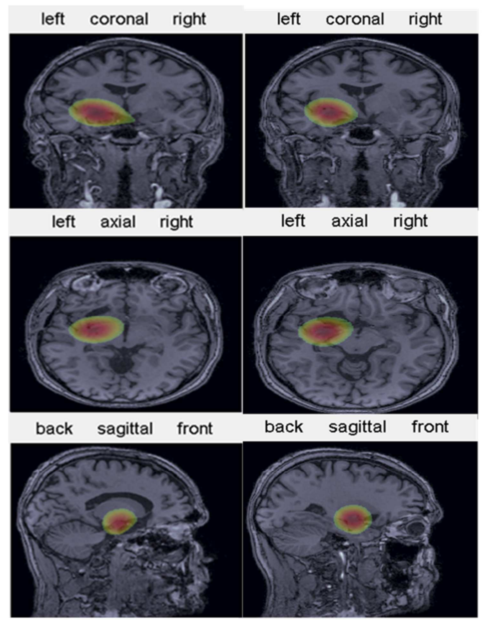

Figure 3. The seizure onset zone for the first seizure of patient 3 is shown. The left column shows the results in three orthogonal slices (coronal, axial and sagittal) for the reference EEG data and the right column shows the results for the EEG data of the wireless prototype.

The main goals of the test were to compare the signal of the wireless prototype with a reference and to check whether the signal quality was good enough for automatic analysis aiming to seizures detection and localization. The seizure detection and localization algorithms developed in a previous work [8] have been applied to both the reference and the wireless prototype data.

IV. RESULTS AND DISCUSSION

Electrical tests on the device gave similar results to the ones reported in [4] for the AE and SN, for both EEG and ETI. The only exception was an increased level of noise higher than 60 nV/$\sqrt{Hz}$ to a value ranging from 80 to 150 nV/$\sqrt{Hz}$ depending on the proximity of channels with the Bluetooth transceiver. The reason for that was discovered in some layout weakness which decreased the immunity of the system to electromagnetic interferences (EMI) generated by the RF transmission. The power consumption for the device was about 350 mW, when acquiring EEG at 600 Hz and ETI at 150 Hz from 48 channels.

A total of about 100 hours of parallel recording have been done on the subjects. One seizure happened during this time and it was correctly recorded by the reference system and by the wireless prototype. Other seizures (a total of 12) happened in periods in which the wireless prototype was not recording (for example during the night)

In the data segments without seizure recordings the false alarm rate has been calculated to compare the two systems. The definition of a false alarm is that the system gives an alarm without the presence of a seizure. The false alarm rate defines the number of false alarms per hour. Algorithm results for the reference and wireless device are reported in TABLE I, which summarizes also the number of seizures recorded by the two systems and detected by the algorithm in each patient. Even if the analysis is not relevant from a statistical point of view, it demonstrates the concept: the false alarm rate of the two systems is very similar. We think that the small differences are due to the fact that the data amount is not exactly the same, because there was a minimal percentage of lost data during the transmission in the case of the wireless prototype and because of the fact that it was stopped during charging periods. Of course, the mean false alarm rate depends on the signal quality of the recorded segments and since they were not exactly the same we have some variations between the two systems (c.f. TABLE I).

For the seizure recorded with both systems we calculated the origin of the rhythmic seizure activity detected by the seizure detector. The results are shown in Figure 3 where the left column shows the results based on the reference EEG and the right column shows the results based on the EEG of the wireless prototype. Thanks to the provided patient individual MRI we were able to use a patient specific head model leading to more accurate estimations which can be visualized in the patient's own MRI. As it can be seen in the source localization results for the onset of the seizure are almost identical for both the systems. The estimated origin of the seizure patterns indicate that the patient probably suffers from left mesial temporal lobe epilepsy.

The only seizure happened during the wireless prototype recording activity was correctly recorded and localized, confirming that the signal quality is good enough to enable automatic analysis of data. Since most of the seizures occurred during the night, when the device was not recording, it is not possible to provide a meaningful evaluation of the sensitivity of the device; a possible solution to enable continuous monitoring is the use of two interchangeable batteries, so to use the second one while the first one is being recharged. On the other hand the choice of recording during the day, when the subject was awake and free to move, allowed verifying the reliability of communication. Despite of the quite high data rate, there was less data loss (about 3%), mainly when the subject exited from the room, outside the range of the receiver. These gaps, didn't affect the automatic data analysis significantly and could be eliminated with the use of a backup non-volatile memory to be used during the out-of-range periods.

V. Conclusions

A high density (up to 64 channels) wireless EEG system has been designed, starting from ASIC based analog front-end, prototyped and tested in a clinical environment. Results showed that with proper compression Bluetooth can support the data rate needed for real-time continuous monitoring of EEG and ETI. Parallel recording of EEG with a reference system commonly used in clinical environment was performed and automatic data analysis for seizure detection and localization provided similar results for the two devices, confirming the reliability of the analog front-end and of the whole system form the point of view of signal quality.

A further development of the work should include a revised system prototyping and packaging. The first goal should be to fully exploit the advantages of active electrodes ASICs, putting them directly on patient head. Furthermore this would allow the reaching of a higher level of integration between SNs (which are currently plugged on the motherboard) and PPN so to optimize the size of the device and to make it really wearable.

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