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Micropower Non-contact EEG Electrode with Active Common-Mode Noise Suppression and Input Capacitance Cancellation

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***Abstract*— A non-contact EEG electrode with input capacitance neutralization and common-mode noise suppression circuits is presented. The coin sized sensor capacitively couples to the scalp without direct contact to the skin. To minimize the effect of signal attenuation and channel gain mismatch, the input capacitance of each sensor is actively neutralized using positive feedback and bootstrapping. Common-mode suppression is achieved through a single conductive sheet to establish a common mode reference. Each sensor electrode provides a differential gain of 60dB. Signals are transmitted in a digital serial daisy-chain directly from a local 16-bit ADC, minimizing the number of wires required to establish a high density EEG sensor network. The micropower electrode consumes only 600μW from a single 3.3V supply.**

I. Introduction

Electroencephalogram (EEG) recording on the scalp offers a non-invasive means to monitor collective and coherent neural activity in the human brain at temporal and spatial scales of interest in various applications ranging from medical diagnostics to brain-computer interfaces [1].

Conventional EEG recording technologies involving gel based electrodes have been well established and are ubiquitous in EEG applications. However, they suffer from several drawbacks arising from the need for extensive preparation and direct electrode contact to the subject's scalp.

As an alternative, capacitive type electrodes that do not require direct contact are attractive since they require a minimal of preparation. Early work [2] [3] have shown efficacy in resolving EEG and ECG type signals via capacitive coupling. Until recently, however, the microelectronics were not available to realize capacitive electrodes competitive with traditional gel-based designs.

Prance et. al [4] [5] demonstrated the use of modern integrated amplifiers to achieve the ultra high input impedances necessary to resolve signals through small coupling capacitances. Other contact-less sensor approaches [6] [7] [8] have successfully met or exceeded the performance of gel electrodes in ECG. Recent contributions have also shown that capacitive electrodes can be used for sensitive EEG recordings when used as part of a hybrid contact sensor [9].

Capacitive electrodes are especially difficult to employ compared to their gel-based counterparts due to the high capacitive source impedance. In ECG applications, this can be somewhat mitigated since large electrode surfaces are permissible [7] [8]. However, for EEG, the combination of small electrode sensing surface (1 to 50pF) and low signal frequencies (1-100Hz) result in extremely high source impedances – on the same order as the input impedance of the electrode's amplifier. Therefore, even small variations in the coupling capacitance and distance between channels can lead to large amounts of distortions due to signal attenuation and the resultant channel-to-channel mismatches.

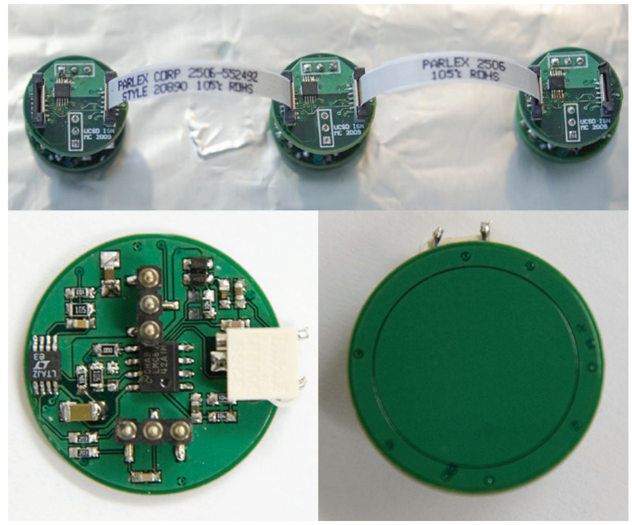

Fig. 1. ADC board with daisy chain. (Bottom Left) Sensor board top surface. (Bottom Right) Sensor board bottom electrode surface.

In this paper, we present an active electrode design for EEG sensing that improves on previous capacitive, contactless electrode designs [10] [11] by actively neutralizing [12] the sensor's input capacitance thereby making the gain of each electrode uniform and constant, irrespective of coupling distance and strength.

II. Electrode Construction

Each sensor (Fig. 1) consists of two small round electrically connected standard printed circuit boards (PCB). The bottom board is the size of a US quarter and contains the sensing plate along with the analog amplifier electronics. The top board, the size of a US nickel, contains the ADC and digital data interface.

The ADC output from each board is a serial data stream which is shifted in a daisy chain [13] from board to board to the end of the chain which connects to a custom USB data acquisition interface. This connection scheme minimizes the

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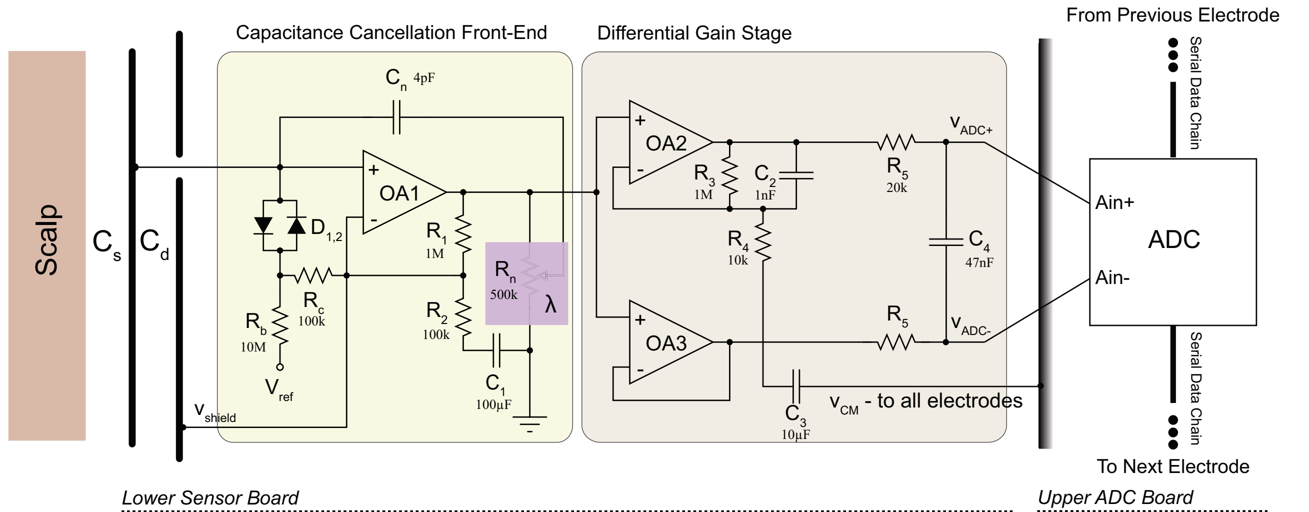

Fig. 2. Schematic of the entire active electrode. The first stage (OA1) performs input capacitance cancellation through C_n and provides a fixed gain of 11. The second stage (OA1, OA2) is connected to a system-wide common-mode line, V_{cm}, and provides a differential gain of 100 along with filtering for the ADC. Data output from the onboard 16-bit ADC is output to a common serial daisy chain.

amount of cabling required across the sensor network, where many electrodes can be operated from a single signal chain, rather than requiring a full cable for each electrode.

Electrical activity from brain are sensed through a $324mm^2$ copper fill insulated by soldermask on the lower board, which is shielded from external noise by the outer copper ring and a solid metal inner plane above the sensing surface. It was found during testing that the nominal maximum coupling capacitance (formed by placing the second plate directly over the bottom soldermask) varies as much as 50% (100pF to 150pF) due to manufacturing tolerances.

The amplification circuits are placed directly on the top surface of the sensor board and outputs a differential analog signal which is digitized by the upper board.

III. Amplifier Circuit

Figure 2 shows the schematic for a single electrode channel consisting of three operational amplifiers split into two stages. The high impedance front-end with the capacitance neutralization circuit is performed by OA1 and provides a gain of 11 followed by a second differential stage [11] with a gain of 100. The LMC6442 operational amplifier was used since it had the lowest current noise ($0.2fA/\sqrt{Hz}$) of all the amplifiers that can operate on a single 3.3V supply. In addition, the power consumption is extremely low – 1μA per channel. Although voltage noise is relatively high in this device the ultra-low current noise makes the amplifier a better choice for recording EEG frequency signals.

A. Front-end Amplifier with Capacitance Cancellation

The sensing capacitor, C_s, is formed by the skin and the plate of the electrode connected to the non-inverting input of OA1. The gain of the first stage is simply set by $A_{v1} = 1 + R_1/R_2$ to 11 and C_1 is a large DC block to ensure the effectiveness of the input capacitance cancellation circuit over all EEG frequencies to avoid a sharp narrow-banding of the gain response as C_s decreases. This front-end provides both impedance conversion and signal gain. Thus, overall electrode sensitivity, noise levels and channel matching is heavily dependent on this front stage.

Designing the input node of the electrode is a challenge since the bias network must not degrade the input impedance of the amplifier or add excessive noise. To avoid the use exotic ultra high value resistors (beyond 1TΩ) to minimize the input noise, the bias network is instead built around two low-leakage anti-parallel diodes, $D_{1,2}$, bootstrapped through, R_b and R_c. The diodes ensure that the DC bias of the input is set to V_{ref} while the bootstrap keeps the dynamical voltage across the diodes near zero. As a result the leakage across the diode is in the tens of femto-amps keeping the noise contribution at a minimum and effective impedance high.

Electrode shielding is accomplished by using the low impedance node of OA1's inverting input to drive a plane above the sensing surface, which forms a capacitance, C_d, between the non-inverting input and, C_s. A well known technique, this method of active shielding [5] [10] does not introduce any extra capacitive loading at the input. However, input capacitances from the amplifier's input C_{in} as well as any stray board level capacitances are still unavoidable, and previous contributions made no active effort to combat this quantity, despite its detrimental effect on gain and channel matching [10] [11] due to the voltage divider formed by C_s and C_{in}.

Consequently, even in electrodes with low input capacitance amplifiers, gain tends to vary significantly [10] as electrode to skin distance is changed. In this design, a neutralization system [12] [14] is added to eliminate residual input capacitances that are not handled by the active shield.

The neutralization circuit employs a potentiometer, R_n, which controls the amount of positive feedback, λ, which is coupled back to the input through the neutralization capacitor, C_n. The overall effect of the neutralization on the input is to make the net input capacitance,

$$C'_{in} = C_{in} - (\lambda - 1)C_n. \quad (1)$$

By carefully adjusting R_n, it is possible to arrive at a value of λ such that the input capacitance is almost completely negated, resulting in an amplifier gain that is essentially invariant with respect to C_s. Although it is possible to over-compensate and create an effectively negative input capacitance, the circuit is stable as long as $C_s + C'_{in} > 0$. Thus even small deviations around the optimal value of λ (for example, the amplifier's input capacitance chances slightly with the input common-mode voltage) are relatively innocuous as the amplifier will remain stable over the range of coupling capacitances encountered in actual usage.

The overall output of the input capacitance neutralizing first stage including noise with the optimal value of λ is approximately,

$$V_{out1} = A_{v1}[v_s + \frac{i_n}{sC_s} + v_n(1 + \frac{C_{in} + C_d + C_n}{C_s})], \quad (2)$$

where i_n is the total current noise arising from the amplifier's bias current and diode leakage and v_n is the equivalent input voltage noise of the amplifier.

From a noise perspective, the SNR with respect to the voltage noise is still set by the ratio of the coupling capacitance versus the sum of all the capacitances at the input node. This highlights the importance of minimizing the amount of coupling from the shield and neutralizing capacitor back to the input (while still maintaining their effectiveness) to minimize the amount of excess noise. Likewise the SNR with respect to current noise is still fundamentally set by the ratio of i_n and C_s, highlighting the importance of low current noise biasing techniques.

B. Differential Gain Stage with Common-Mode Suppression

The second stage of the amplifier establishes a global common mode line, v_{cm}, and provides differential gain over a bandwidth of 1-100Hz. The circuit follows the same design as in [11], where a full analysis is presented. The amplifier OA2 is coupled to every other electrode through v_{cm} which averages the signal over the subject. This common mode signal is used as a reference for differential signal gain as well as active grounding in a driven right leg circuit and subject shielding.

The output presented to the ADC is a filtered differential signal with a gain of 1100, making it possible to resolve the small EEG signals.

IV. Sensor Characterization

A. Input Capacitance Neutralization and Gain

To test the effectiveness of the input capacitance cancellation circuit, R_n was carefully adjusted so that the gain is constant even as the electrode is spaced away from signal generator through stacks of glass slides. Figure 3 shows a plot of the gain versus the sensor separation distance with the neutralization circuit both on and off.

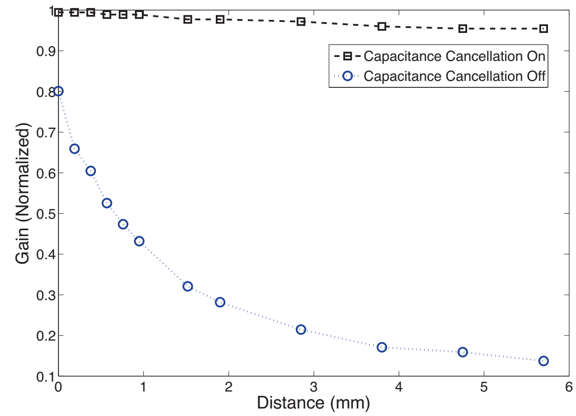

Fig. 3. Plot of mid-band sensor gain versus separation distance to signal source. Gain is normalized where 1 is the maximum possible gain ($C_s = \infty$).

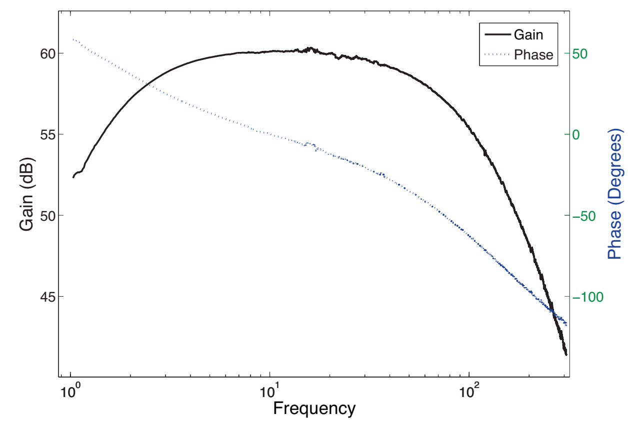

Fig. 4. Sensor differential-mode gain and phase at maximum coupling capacitance.

From this data, it can be seen that even with the maximum coupling strength (sensor plate separated only by soldermask), it is impossible to achieve the full gain of the amplifier without the neutralization circuit. Furthermore, since the sensor PCBs have an inherently large variation ($C_{s,max} \pm 50\%$), large electrode mismatches occur even in the best of cases. Using the capacitance neutralization circuit trims each electrode to the specified maximum gain.

B. Power

Power consumption was measured to be only 600μW per electrode. This includes all three amplifiers, ADC and digital interfacing.

V. Physiological Recordings

A two electrode EEG acquisition system was implemented with the capacitive electrode. As reference, one electrode was placed below the ear, while a second electrode was placed on the forehead. Subject grounding was accomplished with a driven right leg circuit derived from the output of the reference electrode and attached back to the side of the

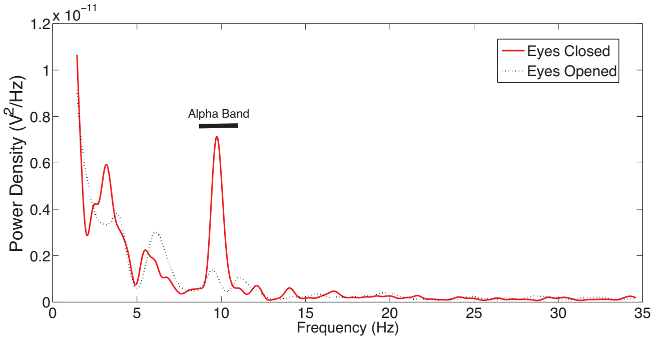

Fig. 5. Power spectrum of recorded EEG signal from the two electrode test for a 10-second period where the subject's eyes were open followed by a 10-second period where the subject's eyes were closed.

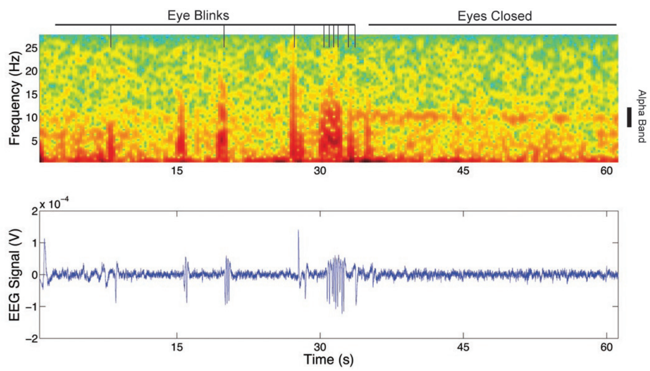

Fig. 6. Spectrogram of recorded EEG signal where the subject blinked, then closed his eyes. Power near 10Hz can be seen after the eyes close indicating the presence of alpha activity.

body with a simple conductive foil. All experiments were conducted in a typical electrical engineering laboratory, and the electrodes were completely unshielded, except for the internal PCB guard plane.

An eye blink and alpha wave experiment is a good first test of the electrode's operation. Figure 6 shows the recorded waveform over an eighty second time period along with the system's spectrogram. Spikes in the recording correspond to the subject's eye blinks. Approximately half way into the segment, power at 10Hz can be observed due to the presence of alpha wave activity when the subject closed his eyes. Likewise, Fig. 5 shows the power spectrum of the two different periods. A clear 10Hz peak is present during the time where the eyes were closed.

The electrode design is also effective for recording high fidelity ECG signals without the need for direct electrical contact (Fig. 7).

VI. Conclusion

The addition of an input capacitance cancellation circuit to the contact-less EEG electrode remedies one of the major short comings encountered in capacitive sensors. By making the effective input capacitance of the sensor essentially zero, the neutralization circuit virtually eliminates gain variation and channel mismatch due to changes in the body to electrode coupling capacitance.

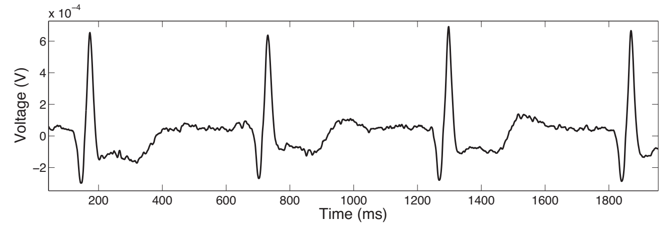

Fig. 7. ECG signal acquired over the subject's chest.

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