Relative Measurement of Differential Electrode Impedance for Contact Monitoring in a Biopotential Amplifier

Sun K. Yoo

Abstract: In this paper, we propose a simple and relative electrode contact monitoring method. By exploiting the power line interference, which is regarded as one of the worst noise sources for bio-potential measurement, the relative difference in electrode impedance can be measured without a current or voltage source. Substantial benefits, including no extra circuit components, no degradation of the body potential driving circuit, and no electrical safety problem, can be achieved using this method. Furthermore, this method can be applied to multi-channel isolated bio-potential measurement systems and home health care devices under a steady measuring environment.

Keywords: Electrode contact monitoring, power line artifact, relative impedance measurement.

1. INTRODUCTION

Electrode contact monitoring is important in the field of biopotential amplifiers [1-6]. Source impedance, skin-to-electrode impedance, and the lead-to-amplifier impedance have a significant effect on the amount of power line artifacts and the CMRR (Common Mode Rejection Ratio) in bio-potential amplifiers [2,4,7,8]. Hence, poor contact of the electrodes to the skin surface and/or broken leads will severely limit the performance of bio-potential amplifiers.

Both current and voltage sources with detection circuitry have been used to measure source impedance [1,3-6]. The contact monitoring circuit incorporated in the bio-potential amplifier increases the number of circuit components, and may degrade the performance of a body potential driving circuit, such as the right leg driven circuit in an ECG, if the oscillating frequency of the current/voltage source signal is not properly chosen [5,7]. Moreover, the application of a source signal to the patient increases the risk of electrical shock if an electrical circuit malfunction occurs.

We regard the aim of contact monitoring as not for the exact measurement of differential source impedance at the differential amplifier inputs, but rather the relative inspection of impedance imbalance to assure the proper attachment of electrodes and the lead wire. Under this constraint, we propose a contact monitoring method by means of measuring the relative impedance differential between electrodes without resorting to any additional current or voltage source. Instead, the displacement current due to the power line interference, which generally exists during patient monitoring, plays a central role as the external current source for impedance measurement without additional circuit components, degradation of the body potential driving circuit, or electrical safety problems.

2. METHODS

The power line coupling capacitances to the human body (Z_p) and the lead wires (Z_a, and Z_b) connected to two separate differential amplifier inputs are the prominent sources of the power line artifacts that contaminate measured biopotential signals.

Displacement current (i_p) due to capacitive coupling between the power line and body (Z_p) produces a voltage drop (V_{Z_3}) at the ground electrode (Z_3), as shown in Fig. 1.

$$V_{Z_3} = i_p \frac{Z_e}{Z_e + Z_s} Z_3, \tag{1}$$

where Z_e, and Z_s are body-to-earth impedance and isolation impedance, respectively. The i_p also produces a voltage drop (V_{Z_t}) at Z_t (the body impedance between electrodes) for the skin impedance. Both V_{Z_3} and V_{Z_t} are associated with

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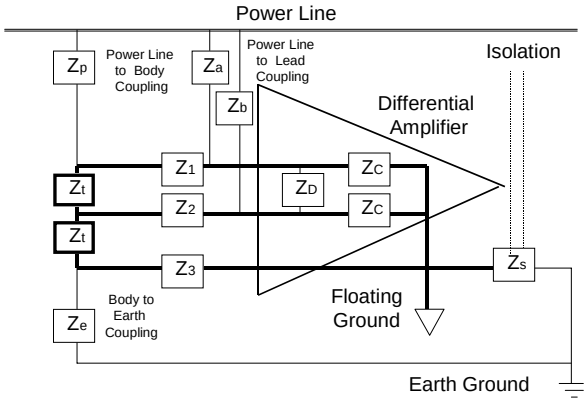


Fig. 1. Circuit model for relative measurement of differential electrode impedance.

differential voltage (V_{Z_D}) between two inputs of the differential amplifier. Hence, V_{Z_t} can be expressed by impedance imbalance due to different electrode contacts (difference between Z_1 and Z_2) in terms of V_{Z_3} and V_{Z_t}. Those circuit components can be simplified by Thévenin equivalent voltage (V_{TH}) and equivalent resistance (R_{TH}) in concatenation with Z_D.

$$V_{TH} = V_{Z_3}\frac{Z_2 - Z_1 - Z_t}{Z_C}, \qquad R_{TH} = Z_1 + Z_2 + Z_t, \tag{2}$$

where $Z_{c,}$ is input impedance of the differential amplifier, and Z_1 and Z_2 are skin-to-electrode impedances. From Thévenin equivalent circuit based on the Pallás-Areny's model [8], the V_{D1} (voltage drop at Z_D) due to Z_p is expressed by

$$V_{D1} = k_1\Delta Z + C_1, \tag{3}$$

where $k_1 = \frac{i_P}{Z_c}\frac{Z_e Z_3}{Z_e + Z_s}$, $C_1 = i_P(Z_t + \frac{Z_e Z_3}{Z_e + Z_s}\frac{Z_t}{Z_c})$.

If we assumed that Z_3 has good contact with the body, and i_p, Z_s, and Z_e change little throughout the measurement, V_{D1} is proportional to the impedance imbalance, $\Delta Z = Z_2 - Z_1$, with the fixed constants k_1 and C_1.

The voltage drop at Z_D due to the lead coupling, V_{D2}, is also represented by the impedance imbalance based on Metting van Rijn's model [7].

$$V_{D2} = k_2\Delta Z + C_2, \tag{4}$$

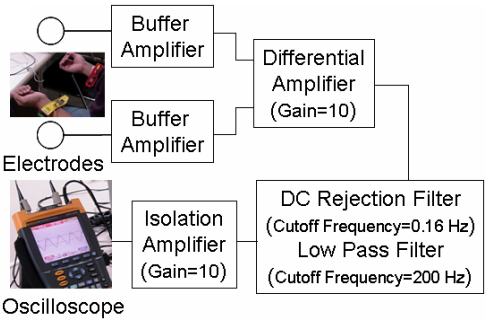
where $k_2 = \frac{V_{Zs}}{Z_s}$, $C_2 = V_{Zs}\frac{Z_1 + Z_2}{2}\frac{Z_a - Z_b}{Z_s Z_s}$, $Z_s = \frac{Z_a + Z_b}{2}$, and V_{Zs} is the isolation mode voltage at Z_s. Similarly, if we assume that Z_p, Z_a, Z_b, Z_1 and Z_2 change little, then k_2, and C_2 will be approximated as a fixed constant. Since half-cell potentials associated with Z_1 and Z_2 can produce an offset voltage due to ionic disturbance at the electrical double layer between the electrode and the skin, then the DC rejection filter, generally inserted at or after the differential amplifier, can remove the offset voltage, which results in the accentuation of the 60Hz components [10]. After offset removal, the combination of equations (1) with (2) will be expressed as a function of ΔZ with k and c being fixed.

$$V_D = V_{D1} + V_{D2} = k\Delta Z + c \tag{5}$$

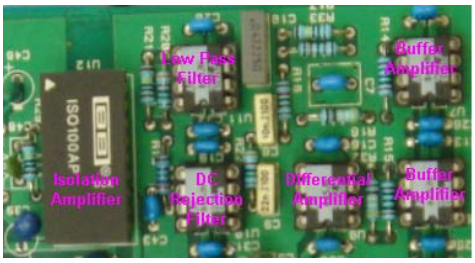
Since V_D is proportional to the impedance imbalance, the power line artifact appearing on the output of the isolation amplifier is a function of impedance imbalance. Conversely, the differential electrode impedance, corresponding to the impedance imbalance, can be relatively measured by observing the power line artifacts, without the use of an additional current or voltage source.

3. RESULTS

In order to validate the relative contact monitoring functionality in a practical biopotential measurement,



(a) The schematic diagram for one channel differential impedance measurement by means of an oscilloscope. The one channel measurement blocks consist of two buffer amplifiers, a differential amplifier, a DC rejection filter, low pass filter and an isolation amplifier.



(b) Implemented prototype board.

Fig. 2. Experimental configuration.

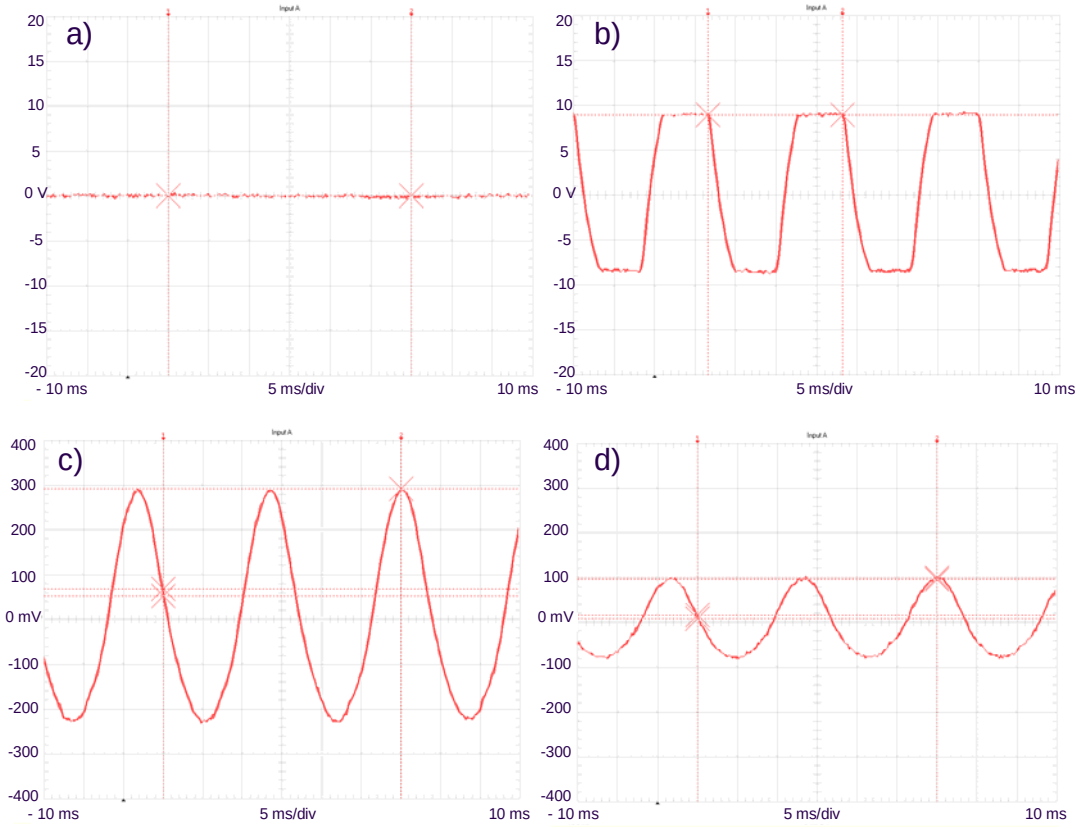


Fig. 3. The 60-Hz output signals typically observed during biopotential measurement are associated with electrode contact conditions. Electrode contact status goes from poor (a) to good condition (d).

a testing board was constructed. The measurement circuit block for each channel consisted of a differential amplifier with a gain of 10, DC rejection filter with a cutoff frequency of 0.16Hz, low pass filter with a cutoff frequency of 200Hz, and an isolation amplifier with unit gain, as shown in Fig. 2(a) and 2(b).

Fig. 3 shows output signals that are typically observed during biopotential measurement. In the case of a broken lead or a poor electrode-skin contact, the high gain and non-ideal electrical characteristic of the differential amplifier cause its output to saturate. As shown in Fig. 3(a), the saturated signal appears similar to the baseline signal due to saturation. If the signal is partially saturated, Fig. 2(b) is observed. If its output does not saturate, the differential contact impedance is proportional to the peak-to-peak amplitude of the power line artifact components as shown in Fig. 2(c) and 2(d). Similarly, ECG (Electrocardiogram) signals with overlapped 60 Hz artifacts with respect to left arm and right arm electrodes were measured under different electrode contact conditions. As shown in Fig. 4(a), 4(b), and 4(c), the amount of 60Hz artifacts caused by impedance imbalance increases as the electrode contact conditions worsen. The measured impedances for Fig. 4(a), 4(b), and 4(c) are 2.6 kΩ, 16.8 kΩ, and 82.5 kΩ, respectively. Highly contaminated 60Hz artifacts due to broken leads prevent ECG signal from being observed as shown in Fig. 4(d).

By adjusting the amount and dryness of electrode jelly and the electrode attachment status, the electrode contact impedance to the skin was artificially changed. After measuring the contact impedance by an external impedance meter (KTMed Co., Korea), the power line

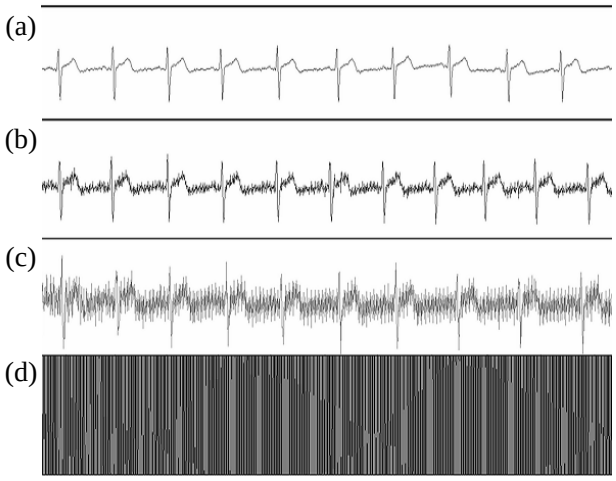


Fig. 4. Measured ECG signals with different contact impedances to the skin surface from good (a) to poor (c), and with broken lead wire (d).

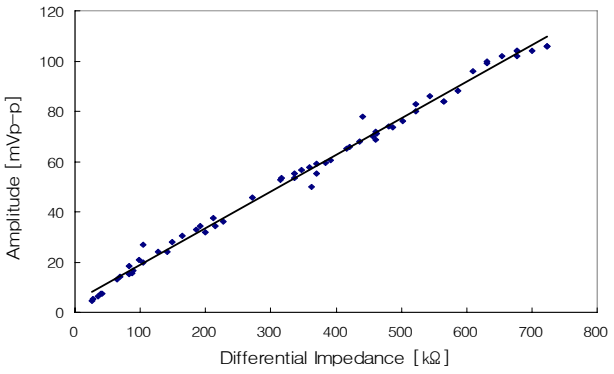


Fig. 5. Peak-to-peak amplitude of the power line artifact component in terms of differential contact impedance.

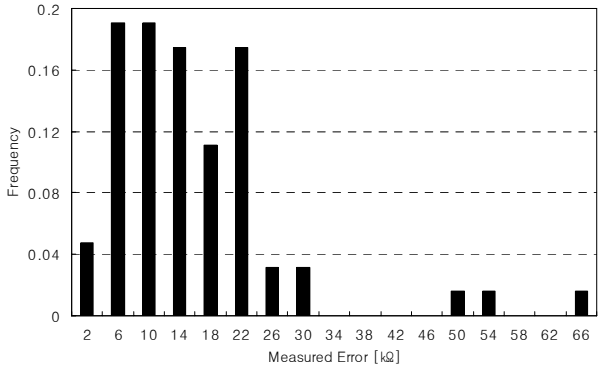


Fig. 6. Measured errors between proposed relative measurement methods and external impedance meters. The mean and standard deviation of the measured errors are 13.9 kΩ, and 11.7 kΩ, respectively.

artifact components were extracted. As shown in Fig. 3, the peak-to-peak amplitude of the power line artifact component is proportional to the electrode contact impedance, which is consistent with (3).

The relative measurement method is compared with the constant-current method of direct impedance measurement using an external impedance meter. Fig. 6 shows the frequency (the number of occurrences/ total number of measurements) in terms of measured error (measured impedance difference between the relative method and the external impedance meter). Before performing an experiment, k and c were estimated by using the least square linear regression method. For 63 measurements, the mean and standard deviation of the measured errors are 13.9 kΩ, and 11.7 kΩ, respectively. Because the contact impedance of 100kΩ, is generally encountered in a normal biopotential measurement [10], most cases (less than 30 kΩ) are manageable for relative contact monitoring. High measurement errors of 50 kΩ, 54 kΩ, and 66 kΩ are caused by body movement (resulting in the changes in estimated k and c) during measurement.

4. DISCUSSION AND CONCLUSION

Poor electrode contact is one of the major reasons for poor performance of measured biosignals [1-6]. However, manual inspection to check electrode contacts can be cumbersome. Intermittent impedance checking can upset the continuous operation of home healthcare devices installed at a fixed location, including the bed and bathtub, where unconscious and uninterrupted measurements are emphasized [9]. If the number of electrodes is small, manual inspection is sufficient. However, there is a tendency to increase the number of channels in modern bio-potential measurement systems, such as brain and cardiac mapping systems [6,7]. The chance of poor electrode contact is more probable as the number of measurement channels increases. For a large number of channels and for home healthcare devices, manual inspection by means of an external lead tester [3] is cumbersome. Moreover, the contact monitoring circuit increases the complexity in designing the amplifier circuit. Hence, a relative measurement method without any extra circuit components can be useful.

In order to determine the warning threshold for poor electrode contact, k in (3) should be measured. However, we cannot obtain k directly in this relative measurement method without the use of impedance measurement circuitry. More importantly, k is not a fixed constant because it depends on the measuring environment, e.g., the power-line voltage level, the distance from the power line to the patient, lead wire arrangement, and circuit board arrangement. Calibration can supplement this relative measurement method. Instead of using k, the warning threshold can be indirectly determined by calibration. The user can select one or more good-quality signal channels as the calibration source under the assumptions that the selected electrodes have good contacts. In other words, the user intentionally maintains good electrode contact to form the calibration source. Based on the averaged peak-to-peak level of power line artifact components extracted from the calibration source, the threshold can be determined to satisfy the assumption and approximation of equation (3).

Special caution is required to apply this method to critical situations, such as EEG measurement in deep hypothermia surgery [11]. When a patient's body temperature is sufficiently low, EEG signals will be almost undetectable, therefore, discrimination between the saturated power line artifact signals (noted in Fig. 3) and the small biopotential signal might be difficult. The saturated artifact signal represents as output noise due to the loss of signal strength. Particularly, the ratio between the output noise and bio-signal level is predominantly influenced by the noise at the buffer amplifier (refer to Fig. 2(a)). The noise level at the unity-gain buffer amplifier can

be approximately modeled by noise voltage, noise current and thermal noise across the resister [12]. Since operational amplifiers are generally employed as buffer amplifiers in a bio-potential amplifier [1,2,5-9], noise level for two buffer amplifiers can be approximately $300\,nV/\sqrt{Hz}$ [12]. Hence, total noise up to 40 Hz associated with EEG signal bandwidth is about 1.9 uV $(=300nV\times\sqrt{40})$, which is significant for low amplitude EEG signal measurements down to 1uV.

This relative measurement method is well suited for relatively constant measuring environments. However, it has some drawbacks. First of all, it cannot provide a direct impedance measurement. Thus, it cannot be applied to measure precise electrode contact impedance, but can be applied to measure relative contact monitoring. Secondly, it is not well suited for measurements taken in an environment prone to sudden movements. Since power line coupling capacitances to the human body and lead wires depend on the area of the circuit board and lead wires, and their orientation to the power line source, movement during measurement can cause changes in power line coupling capacitances (corresponding to changes in k). The changed k requires recalibration of the system. Thus, the relative measurement method cannot be used in a high motion environment, such as during exercise, because of the high variability of the constant k. Finally, good contact for the ground electrode (Z_3) is a prerequisite to confine 60Hz artifacts to differential input impedance imbalances. Moreover, 60Hz artifacts associated with ground electrodes are related to all channels. A poorly-contacted ground electrode to the skin or a broken lead for the ground electrode should be carefully avoided to maintain good contact and therefore quality measurements throughout the duration of the experiment.

In this paper, we propose a simple and relative electrode contact monitoring method. By exploiting the power line interference, which is regarded as one of the worst noise sources used in biopotential measurement, the relative differential contact impedance between electrodes can be measured without a current or voltage source. Substantial benefits, including no additional circuit components, no degradation of the body potential driving circuit, and no electrical safety problem, can be achieved. This method can be appropriately applied to multi-channel isolated biopotential measurement systems and home health care devices under a steady measuring environment.

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