Communications

The Isolation Mode Rejection Ratio in Bioelectric Amplifiers

A. C. Metting van Rijn, A. Peper, and C. A. Grimbergen

Abstract—**Galvanic isolation of a patient during a bioelectric recording is necessary to ensure the safety of the patient. In a typical measurement situation high interference voltages may be present across the isolation barrier. In this paper the necessity of a very high isolation mode rejection ratio—the ability of the amplifier to suppress feed-through from voltages across the isolation barrier to the output— is argued and a design of a multichannel amplifier with an isolation mode rejection ratio of 160 dB is described.**

I. Introduction

The safety regulations in medical care prescribe maximum leakage currents in case a patient should touch ground or main power during a bioelectric recording [1]. Although the regulations vary between different countries, a sufficient galvanic isolation and an isolation capacitance less than 800 pF are usually demanded (maximum leakage current is 50 μA_{rms} when the patient touches the line voltage of 220 V_{rms}, 50 Hz). An isolation capacitance of less than 300 pF will ensure that the connection of a recording device will not significantly alter the situation of the patient, the normal body capacitance by ground also being in this order [2].

II. Origin of Isolation Mode Voltage

A typical measurement situation with all relevant capacitances is shown in Fig. 1 (switch opened). The capacitance between the body and main power (C_{body}) causes a current i_{pow} to flow through the body to ground. Part of this current (i_{amp}) flows through the neutral electrode (Z_{rl}, the right leg electrode in ECG recordings) to the isolated common of the amplifier and via the isolation capacitance (Z_{iso}) to ground. Consequently, a voltage across the isolation barrier will exist: the isolation mode voltage (v_{im}) [3].

In every isolation amplifier, a certain portion of the isolation mode voltage is fed-through to the output. The ability of an amplifier to suppress feed-through of isolation mode voltage is quantified in the isolation mode rejection ratio [4]:

$$\text{IMRR} = 20 \log \left(\frac{v_{im} \cdot A}{v_{out}} \right) \tag{1}$$

where

- v_{im} is the isolation mode voltage
- v_{out} is the amplifier output signal due to feed-through of v_{im}
- A is the overall gain of the amplifier.

In this paper two ways of preventing an interference output signal resulting from isolation mode voltages are described. One approach is to reduce the actual voltage across the isolation capacitance with an additional circuit. A better approach, dealt with in Section IV, is to design the amplifier in such a way that the feed-through of high isolation mode voltages remains negligible.

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III. Reduction of Isolation Mode Voltage

The magnitude of the isolation mode voltage is proportional to the interference current through the isolation barrier. The current through the isolation barrier (i_{iso}) can be made much smaller than the current from the patient to the isolated amplifier common (i_{amp}) with the circuit pictured in the inlay of Fig. 1 (switch closed) [5]:

$$i_{iso} = i_{amp} \left(\frac{R_1}{Z_{iso}(1 + A) + R_1} \right) \approx i_{pow} \frac{R_1}{Z_{iso} \cdot A} \tag{2}$$

where A is the open loop gain of the operational amplifier (typically 10^4 to 10^5 at 50 Hz).

Large resistors R_1 and R_2 limit the maximum leakage current through the patient (for example, resistors larger than 40 MΩ for a maximum leakage current of 50 μA_{rms}). The main drawback of the circuit is that the amount of interference current (i_{amp}) which can be handled in normal operation is limited to $i_{max} = v_{max}/R_1$ (v_{max} is the maximum output swing of the operational amplifier). Consequently, the chosen value of R_1 is a compromise between safety on the one hand and isolation mode voltage reduction and current handling capability on the other. Some improvement can be achieved by increasing the output voltage swing of the amplifier, for instance, by adding an extra transistor output stage with a high supply voltage [6], [7].

In cases were safety regulations are not very severe, the circuit can be valuable because it is easily added to a system with an insufficient isolation mode rejection ratio whereas the improvement can be substantial. For example, when C_{pow} is 3 pF (i_{pow} is 200 nA_{rms}) and a C_{body} and C_{iso} both 300 pF, the circuit (with R_1 = 10 MΩ and a typical operational amplifier) reduces the isolation mode voltage from approximately 3 V_{p-p} to 0.3 mV_{p-p}. Note that with R_1 larger than 50 MΩ the circuit would not have been able to function in the typical measurement situation just mentioned.

IV. Improvement of the Isolation Mode Rejection Ratio

A preferable way to reduce the influence of isolation mode voltage is to increase the isolation mode rejection ratio of the amplifier. High isolation mode voltages are present in a situation where C_{pow} is large (the patient is situated near power lines), C_{body} is small (no big grounded objects near the patient), and C_{iso} is small. For example, C_{pow} is 30 pF and C_{body} and C_{iso} both 100 pF leads to an isolation mode voltage of 100 V_{p-p}. The resulting interference signal should be lower than the amplifier noise level. The equivalent input noise of a typical biomedical amplifier is in the order of 3 μV_{p-p} (approx. 0.5 μV_{rms}) in a bandwidth 0.1–500 Hz. Consequently, an isolation mode rejection ratio of 150 dB (at 50 Hz) is required [see (1)].

Commercially available isolation amplifiers (with unity gain) provide an isolation mode rejection ratio of approximately 100 dB at 50 Hz. It follows from (1) that the isolation mode rejection ratio

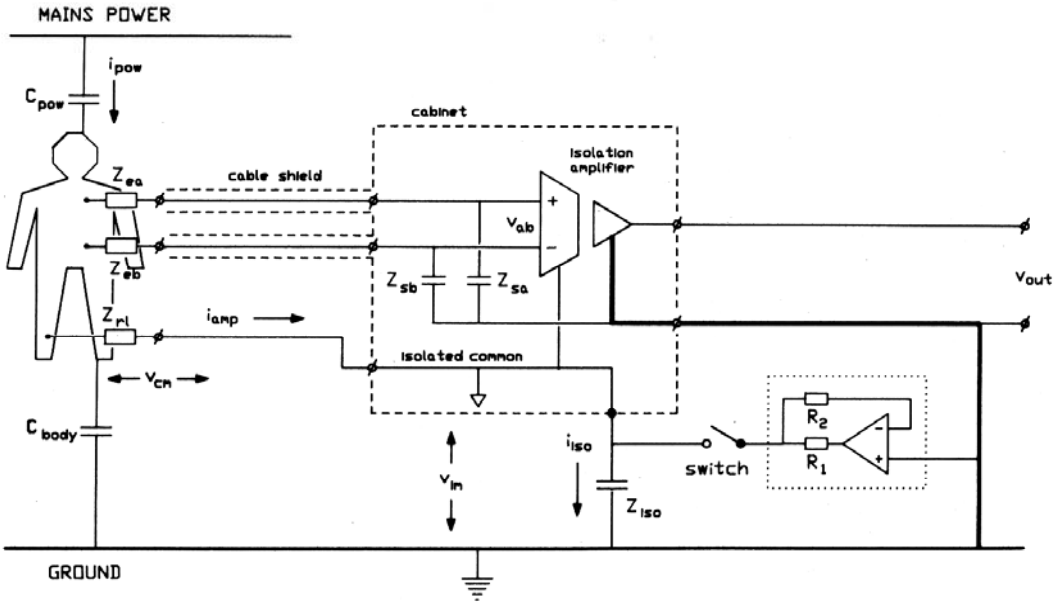

Fig. 1. Schematic diagram of an isolated biomedical recording. With the switch opened i_{iso} is equal to i_{amp}. With the switch closed, the additional circuit makes i_{iso} much smaller than i_{amp} and consequently the isolation mode voltage (v_{im}) is reduced.

can be increased with a preamplifier:

overall IMRR = IMRR of iso amp + gain of preamp (in dB)

The magnitude of the input signals defines the maximum gain of the preamplifier. With ECG measurements—the required input range being approximately 5 mV_{p-p}—the maximum gain is approximately 60 dB and an isolation mode rejection ratio of 160 dB seems obtainable. Note that a dc rejection circuit is essential to prevent saturation of a high-gain amplifier because the differences in electrode offset voltages produce a dc input signal of several tens of millivolts typically.

If the isolation mode rejection ratio is sufficiently high, the isolation mode voltage can still be the cause of interference as differences in electrode-skin impedances and/or stray capacitances convert the isolation mode voltage to a differential amplifier input voltage (v_{ab} in Fig. 1):

$$v_{ab} = v_{im}\left(\frac{Z_{iso} + Z_{rl}}{Z_{iso}}\right)\left(\frac{Z_{sa}}{Z_{ea} + Z_{sa}} - \frac{Z_{sb}}{Z_{eb} + Z_{sb}}\right) \quad (3)$$

where

Z_{iso} is the impedance of the isolation capacitance
Z_{rl} is the electrode–skin impedance of the neutral electrode
Z_{ea}, Z_{eb} are electrode–skin impedances of the measuring electrodes
Z_{sa}, Z_{sb} are impedances of stray capacitances.

It is instructive to rewrite this equation assuming that Z_{iso} is much larger than Z_{rl}, and that the impedances of the stray capacitances are much larger than the electrodes–skin impedances:

$$v_{ab} = v_{im} \cdot \frac{Z_e}{Z_s} \cdot \left(\frac{\Delta Z_e}{Z_e} + \frac{\Delta Z_s}{Z_s}\right) \quad (4)$$

where

$$Z_e = \tfrac{1}{2}(Z_{ea} + Z_{eb});\ Z_s = \tfrac{1}{2}(Z_{sa} + Z_{sb})$$
$$\Delta Z_e = Z_{eb} - Z_{ea};\ \Delta Z_s = Z_{sa} - Z_{sb}.$$

The influence of the isolation mode voltage appears to depend on the stray impedances and their relative differences and on the (with every recording different) electrode–skin impedances. Now consider a relative difference in electrode–skin impedances of 50% and a mean electrode–skin impedance of 100 kΩ at 50 Hz (a "bad" but not unusual measurement situation). It follows from (4) that with equal stray capacitances as small as 0.1 pF—a typical value for the capacitance between unshielded input wires and ground—the isolation mode voltage will produce an input interference voltage of $v_{ab} = (1.5 \times 10^{-6} \cdot v_{im})$, which is equivalent to an (insufficient) isolation mode rejection ratio of 116 dB.

However, even with the use of shielded input wires, small stray capacitances are easily introduced if the input stage and the grounded output stage are located in the same cabinet. It follows from (4) that with the aforementioned electrode impedances equal stray capacitances smaller than 0.002 pF are required to reduce interference to a level equivalent with an isolation mode rejection ratio of 150 dB.

V. Design of an Eight-Channel ECG Amplifier

We developed an eight-channel ECG amplifier with an isolated low-noise input section and eight separate analog output signals. A circuit diagram (one channel) is given in Fig. 2 and the specifications are listed in Table I.

Isolation is accomplished with optically coupled isolation amplifiers[1] (Burr-Brown 3650) offering an isolation mode rejection of 100 dB at 50 Hz (with unity gain). Power supply of the isolated section is by two maintenance-free lead acid batteries (12 V, capacity 2.6 Ah each; maximum continuous operation time is approximately 10 h).

A two-stage preamplifier based on low-noise FET operational amplifiers (OPA 2111) precedes the isolation amplifier. FET operational amplifiers are used because of their low current noise, resulting in lower noise than achievable with bipolar operational amplifiers in measurements with source resistances higher than approximately 30 kΩ. One of the applications of the amplifier is the recording of His-bundle activity at the body surface [8], [9]. A

[1]Optically coupled isolation amplifiers are easy to apply in multichannel applications because no synchronization of carrier frequencies is needed as is the case with inductively or capacitively coupled isolation amplifiers.

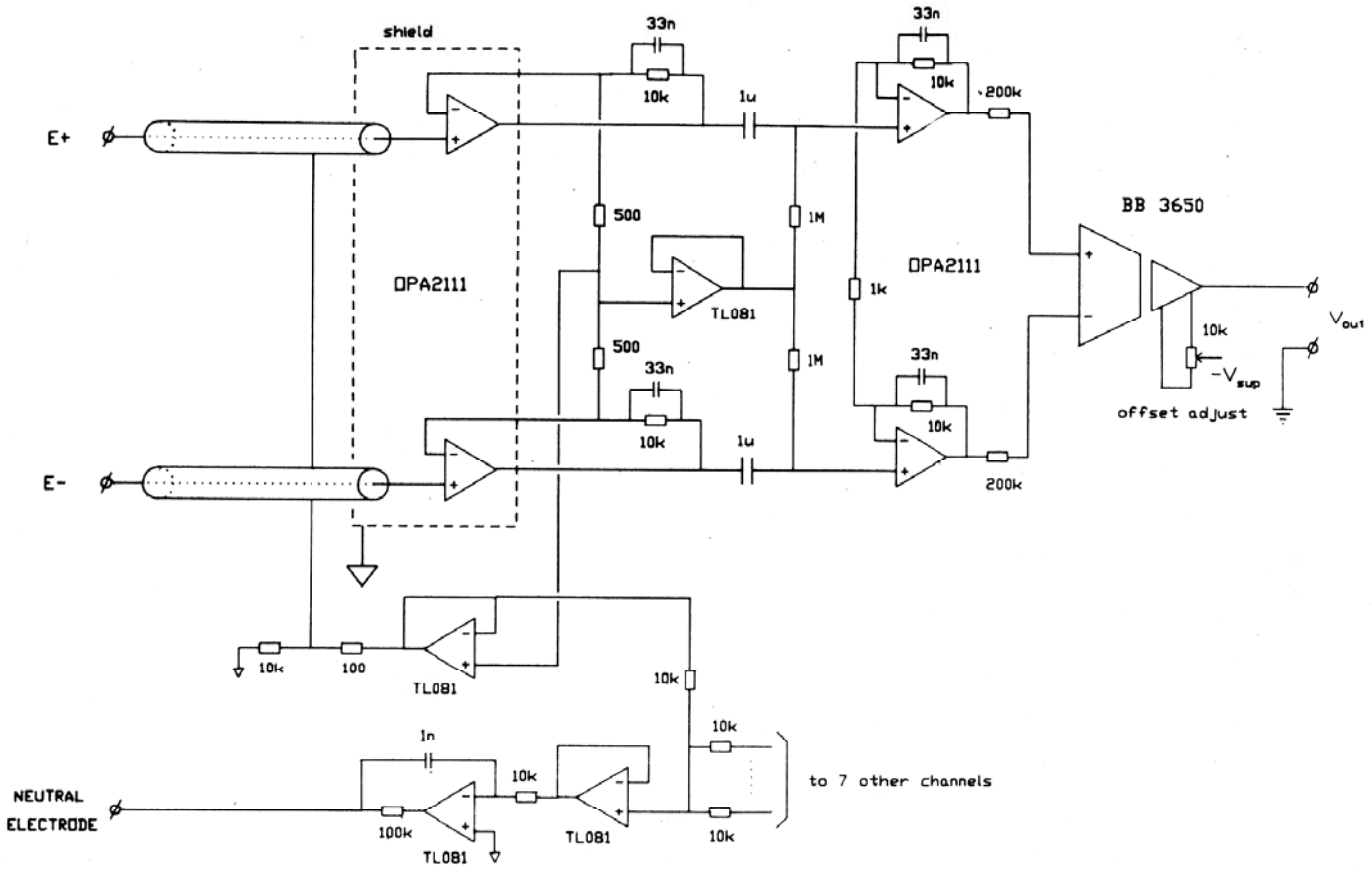

Fig. 2. Schematic of the isolation amplifier (one channel). The shield around the input operational amplifiers is indicated with a dashed line.

TABLE I
AMPLIFIER SPECIFICATIONS

• equivalent input noise voltage (μV_{rms}, 0.1–500 Hz):	0.5
• equivalent input noise current (pA_{rms}, 0.1–500 Hz):	0.03
• bandwidth (+0, −3 dB) (Hz):	0.16–500
• differential mode dc input range (mV):	1000
• differential mode ac input range (mV_{p-p}):	20
• common mode input range (V_{p-p}):	10
• input bias current (nA per input):	0.01
• gain:	1000
• common mode input impedance[1] (MΩ at 50 Hz):	200
• differential mode input impedance[1] (MΩ at 50 Hz):	15
• common mode rejection ratio[2,3] (dB at 50 Hz):	120
• isolation mode rejection ratio[2] (dB at 50 Hz):	160
• maximum leakage current (μA_{rms})[4]:	20
• power consumption (W, eight channels)	2.8

[1]Measured with guarded input leads, cable capacitance is 200 pF.
[2]With a mean source impedance of 100 kΩ and a relative difference in source impedances of 50%.
[3]Measured with the patient connected to the driven right leg circuit.
[4]Eight channels connected, isolation mode voltage is 220 V_{rms}, 50 Hz.

major portion of these recordings is to be performed with neonatals, where we noted relatively high electrode-skin impedances (mean value approximately 100 kΩ).

The low-frequency common mode rejection ratio is made independent of imbalances between the two high-pass filters (required for dc suppression) by connecting the resistors of the filters to a low-impedance version of the common mode signal (Kuiper, personal communication). A guarding circuit [10] ensures a high common mode input impedance with shielded input leads and a driven right leg circuit [11] reduces the common mode voltage by 50 dB at 50 Hz.

Overall gain of the amplifier circuit is 60 dB and an isolation mode rejection ratio of 160 dB was readily achieved with balanced source impedances. However, proper design of the cabinet proved to be essential to prevent interference with unbalanced source impedances. The input leads of the operational amplifiers in the first amplifier stage are mounted directly to the input connectors and the complete input section is shielded with the shield connected to the isolated common. These construction details, combined with guarded input wires, ensure minimum stray capacitances between the amplifier inputs and ground. As a result, the interference level with a mean source impedance of 100 kΩ and a relative difference in source impedances of 50% was equivalent to an isolation mode rejection ratio of 160 dB.

VI. DISCUSSION

It was shown in this paper that an eight-channel isolation amplifier with a very good isolation mode rejection ratio can be built with commercially available components, provided that special attention is paid to the shielding of the amplifier inputs. If a system with more than eight channels is built, the total isolation capacitance of the parallel isolation amplifiers may become unacceptably large. The number of isolation amplifiers can be reduced by multiplexing, i.e., several channels are handled by a common isolation amplifier.

The most promising technique for a multichannel (>20) measurement system is the digital transfer of information. A system with a multiplexer and A/D converter and with serial data transmission through an optical fiber can offer minimum isolation capacitance (the distance between isolated and grounded sections can be large), minimum stray capacitances from amplifier inputs to the grounded section (the isolated section and grounded section can be in different shielded cabinets) and maximum isolation mode rejection (only extremely large interference signals can corrupt the digital information).

The main problem of such an optically coupled system is the power supply of the isolated section as the addition of a dc–dc converter would jeopardize the favorable points just mentioned. Fortunately, power supply with small-sized batteries becomes more and more a realistic alternative with modern low-power components (a multichannel amplifier, multiplexer, A/D converter and optical transmitter with a combined power consumption less than 100 mW are needed). It was shown by our group that multichannel low-power biomedical amplifiers with very good specifications can be constructed [12]. Low-power CMOS A/D converters and multiplexers became commercially available over the last years. However, the relatively large power consumption of the optical transmitters (LED or LASER) remains a problem.

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