Technical note

Low-power, low-noise instrumentation amplifier for physiological signals

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1 Introduction

For a portable multichannel biomedical telemetric system intended to be used for EEG and ECG signals, low-noise, low-power instrumentation amplifiers were required. In general, the design of instrumentation amplifiers for physiological signals is based on the classical configuration with three amplifiers. The main properties of this type of circuits are high input impedance, high common mode rejection ratio CMRR and simple gain control (Tobey *et al.*, 1971; Stott and Weller, 1976; Webster, 1978).

The power consumption of these circuits depends on the circuit parameters of the operational amplifiers (static power consumption) and on the value of the discrete components employed (dynamic power consumption). When operational amplifiers with current programmable characteristics are used the power consumption can be adjusted by means of the set current I_{set}, upon which also the gain bandwidth product GBP and the input noise depends. Adjustment of the GBP by the set current can also be employed to control the high-frequency cutoff necessary in multichannel systems (Radhakrishna Rao and Srinivasan, 1974). However, the GBP is proportional to the set current whereas the input noise voltage is inversely proportional to the square root of this current, and so a compromise between these two parameters has to be made.

In this study use will be made of thick-film technology to acquire the small dimensions essential for a portable device. The choice of thick-film technology inflicts certain restrictions of a technological nature (Sergent, 1981*a*; *b*). To test the usefulness of the classical design for our purpose, an instrumentation amplifier employing current programmable operational amplifiers was built and tested. However, the equivalent input noise voltage of this design did not meet the required upper limit of 0·6 μV RMS* at a power consumption of 150 μW (low enough for implantable systems).

Therefore a different type of physiological amplifier was developed. This circuit employs a dual matched transistor pair input stage and current programmable operational amplifiers (Wong and Ott, 1976; Graeme, 1977; Netzer, 1981).

The circuit has been built and tested using normal electronic components and integrated circuits in standard packages. A thick-film version of this circuit is under development.

2 Circuit description A

Classical design

The realisation of the physiological amplifier derived from the classical configuration with three amplifiers is shown in Fig. 1. DC suppression is effected by feeding the output signal back to the input of A_3 via the integrating network formed by R_4, R_5, R_6, C_f and A_4 (Graeme, 1973). The cutoff frequency of the high-pass filter obtained this way (0·16 Hz) depends on the time constant of the integrator. This circuit is an alternative for passive high-pass filtering which requires too high a capacitor value to implement in thick-film technology. In this circuit, the maximum input DC voltage which can be suppressed (DC suppression range) is equal to the supply voltage divided by the amplification factor. High-frequency cutoff is obtained by the filter defined by R_1 and C_1 and the amplifier poles of A_1, A_2 and A_3.

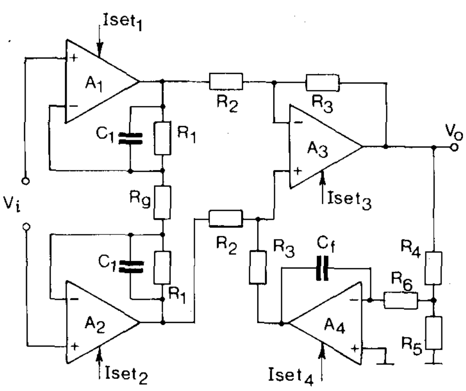

Fig. 1 *Circuit of the instrumentation amplifier for physiological signals derived from the classical three operational amplifiers configuration*
$R_1, R_3, R_4, R_6 = 500\,k\Omega$ $C_1 = 1{\cdot}6\,nF$
$R_2, R_5 = 100\,k\Omega$ $C_f = 280\,nF$
$R_g = 50\,k\Omega$
$A_1, A_2, A_3, A_4 = 1/4$ *LM346* (*National Semiconductor*)
$I_{set1}, I_{set2} = 100\,nA$
$I_{set3}, I_{set4} = 50\,nA$
supply voltage $= \pm 5\,V$

The amplifier poles are related to the GBP, which can be adjusted by the set current I_{set}. The set current also defines the amplifier power consumption and the DC bias current level in the differential input stages of the operational

* *As a good approximation, common electrical noise lies within plus or minus three times the root-mean-square (RMS) value of the noise wave. The peak-to-peak voltage is less than six times the RMS for 99·7 per cent of the time (Motchenbacher and Fitchen, 1973).*

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amplifiers, which in turn is responsible for the input noise level. With the commercially available current programmable operational amplifiers the desired low level of input noise in this circuit can only be achieved by increasing the set current to a level at which the power consumption will not be acceptable.

3 Circuit description B

Improved low-power, low-noise physiological amplifier

In Fig. 2 the circuit configuration of the improved physiological amplifier is shown.

Basically, the design involves two cascade-connected differential transistor stages (WONG and OTT, 1976; GRAEME, 1977), and a current programmable operational amplifier which provides feedback and a single-ended output.

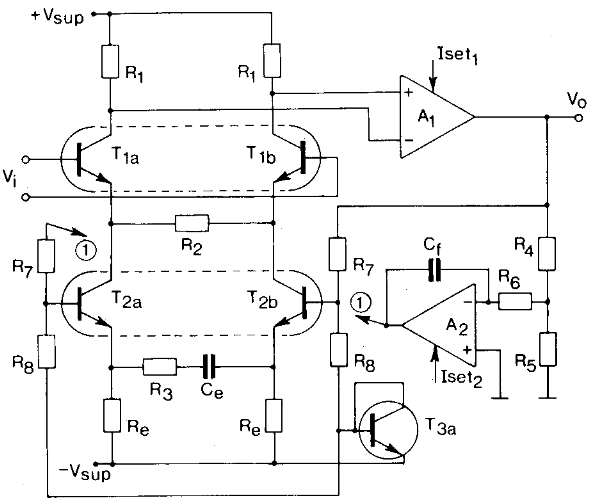

Fig. 2 *Circuit B, the improved low-power, low-noise instrumentation amplifier for physiological signals*
$R_1, R_4, R_6, R_7, R_e = 1\,M\Omega$ $\quad C_e = 220\,pF$
$R_2 = 100\,k\Omega$ $\quad C_f = 150\,nF$
$R_3 = 500\,k\Omega$
$R_5, R_8 = 250\,k\Omega$
$A_1, A_2 = LM4250$ (*National Semiconductor*)
$T_1, T_2, T_3 = LM394$ (*National Semiconductor*)
$I_{set\,1}, I_{set\,2} = 50\,nA$
supply voltage $(\pm V_{sup}) = \pm 4{\cdot}5\,V$

The differential input amplifier is formed by the transistor pair T_{1_a} and T_{1_b}. The transistors T_{2_a} and T_{2_b}, current sources for T_{1_a} and T_{1_b}, are part of the feedback circuit. T_{2_b} is controlled directly by V_0. T_{2_a} receives the output signal via the integrating network, which is realised with A_2. DC suppression is obtained in the same manner as described for circuit A.

The amplifier pole of A_1 and the network formed by R_e and C_e provide high-frequency cutoff. R_3 has been inserted to provide a zero in the transfer function to improve stability in the higher-frequency region. The network formed by R_7, R_8, R_e and T_{3_a} defines the input stage DC biasing and, together with R_2, the amplifier voltage gain.

4 Discussion and results

In this paper, two low-power instrumentation amplifiers are discussed. Circuit A is based on the classical three-amplifier design, whereas in circuit B a dual matched transistor pair is used in the input stage. Circuit B is superior to circuit A with respect to critical parameters such as equivalent input noise voltage (see Fig. 4), CMRR and power consumption* (see Table 1). Another advantage of circuit B is the independence of the power consumption from the input signal DC level. (In circuit A a common mode input direct voltage results in a current increase in both R_2 and R_3.)

In circuit A important parameters such as equivalent input noise voltage (see Fig. 3) and power consumption may be improved by the application of other types of operational amplifiers. When, for example, the current programmable operational amplifier LM 4250 is employed, the equivalent input noise voltage of the circuit might be lowered to $3\,\mu$V RMS (bandwidth: 0·16–200 Hz) and the power consumption to $50\,\mu$W. However, when use is made of commercially available current programmable operational amplifiers (like the LM 4250), it still is not possible to obtain the specifications required with respect to input noise voltage ($0{\cdot}6\,\mu$V RMS) and power consumption (NETZER, 1981).

Owing to the extended low-frequency range of the circuit (low-frequency cutoff point is 0·16 Hz) the $1/f$ noise properties of the input transistors are of utmost importance to its noise performance. In circuit B a transistor pair with optimum properties in this respect can be chosen (see Figs. 2 and 4) (NETZER, 1981).

The rather limited DC suppression range of the circuit can be greatly improved by changing the ratio of the resistors in the transistor input stage, at the cost of the noise figure or the power consumption. Approximately 75 per cent of the power consumption of the circuit occurs in resistors R_7 and R_8 (Fig. 2). When the value of these resistors are increased a decrease of the total power consumption may result, at the expense of a slight increase in input noise.

** Recently, National Semiconductor announced a monolithic high-performance amplifier (LM163), based on the same principles. However, power consumption of this integrated circuit is prohibitive for our application.*

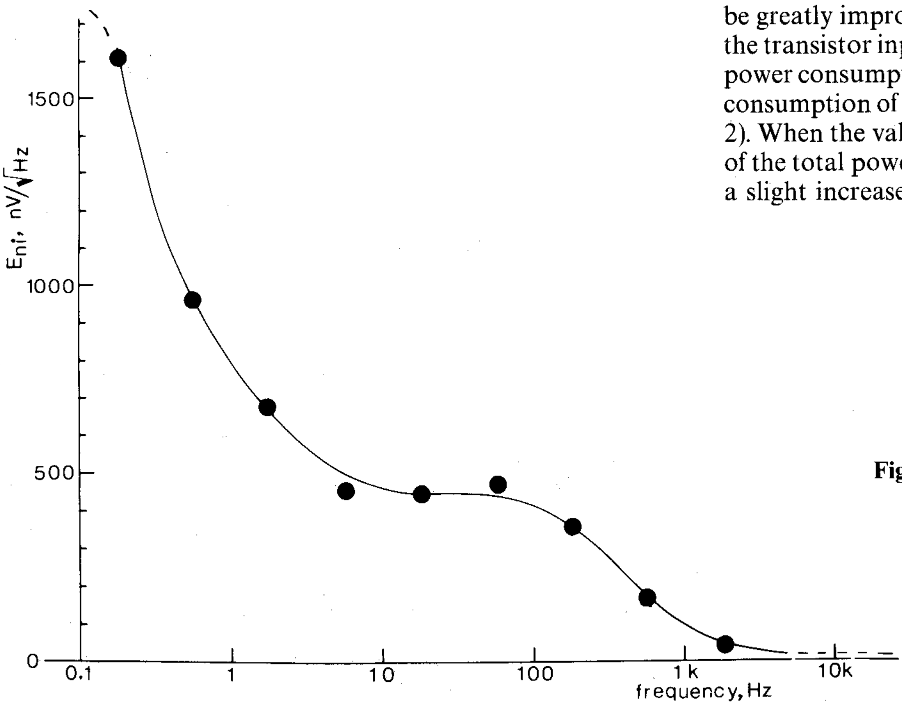

Fig. 3 *Equivalent input noise voltage as a function of frequency for circuit A (see Fig. 1) (measurements with both inputs connected in common via* $10\,k\Omega$*)*

Table 1. Pursued specifications and measured performance characteristics

Parameter	Symbol	Demand value min/max	Circuit A	Circuit B	Unit
Differential input impedance	Z_{id}	>10	>10	>10	MΩ
Input bias current	I_{bias}	<10	<5	<6	nA
Equivalent input noise voltage ($R_s = 10\,k\Omega$, bandwidth = 0·1–316 Hz)	Eneq	<0·6	≈ 7	≈ 0·85	μV RMS
Common mode rejection ratio ($f = 50$ Hz)	CMRR	>60	>60	>85	dB
Voltage gain	A_v	>40·0	40·8	40·3	dB
Harmonic distortion	D	<1·0%	<0·5%	<0·5%	—
DC input range	DCIR	±50	±40	+40/−30	mV
Bandwidth	BW	0·16–200	0·16–200	0·16–200	Hz
Power consumption	PC	<150	≈150	≈100	μW

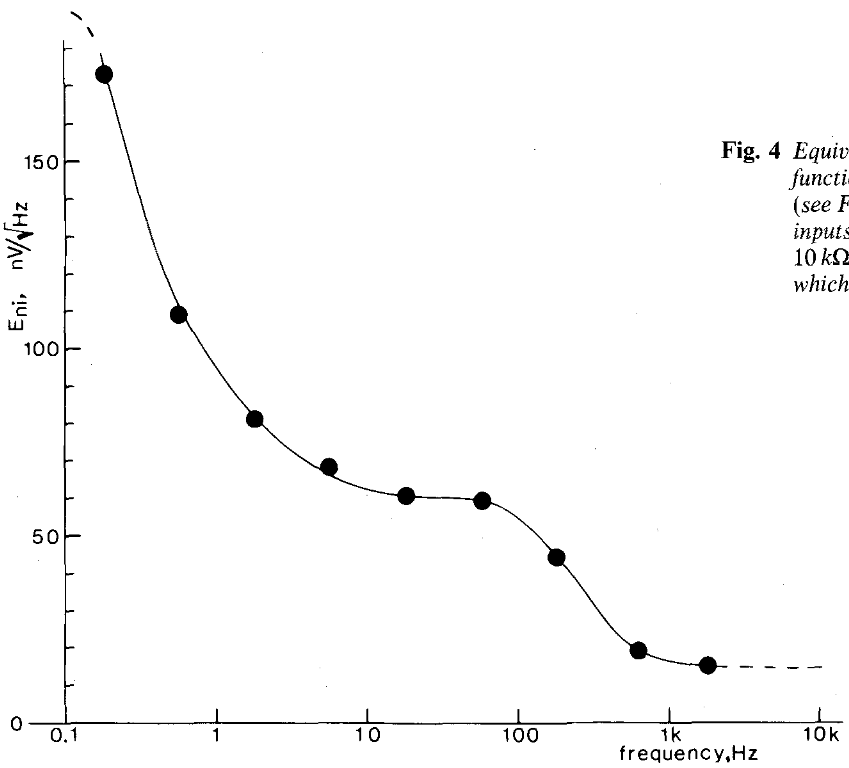

Fig. 4 *Equivalent input noise voltage as a function of frequency for circuit B (see Fig. 2) (measurements with both inputs connected in common via 10 kΩ). Note the noise voltage scale, which differs from Fig. 3*

The bandwidth of the total circuit can be changed by changing passive components only. (The high-frequency range depends on $I_{set\,1}$, R_e and C_e, the low-frequency range depends on R_4, R_5, R_6 and C_f.) Owing to this property, the circuit can be adapted to the measurement of many physiological signals, providing an optimal low-power circuit.

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