VOLTAGE CONTROLLED CURRENT SOURCES FOR BIOIMPEDANCE MEASUREMENT

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Abstract. The paper deals with the requirements and the practical implementation of the voltage controlled current sources used in bioimpedance measuring systems. The importance of the output impedance of the current source for the accuracy of the measurement is described and its minimal value for the predetermined accuracy of the system is calculated. The influence of the electrode/skin contact impedance and the stray capacitances on the performance of the measurement is discussed. In the second part of the paper some practical implementations of the current sources, used in bioimpedance measuring systems, are described. An alternative approach for realization of current source, containing measuring channel for the injected current is proposed.

Key words: Voltage controlled current source (VCCS), bioimpedance measurement, output impedance, stray capacitance.

To measure an unknown impedance we can use one of the two possible approaches. Either we must apply a known voltage across the object and then measure the current through it, or we must inject a known current into that object and measure the voltage across it. In a system where we are measuring the impedance of a patient, so called bioimpedance, an unknown and varying contact impedance exists between the electrode and the patient's tissue. This unknown contact impedance will affect negligibly a good current source which has high output impedance, while it will degrade the operation of a voltage source which has low output impedance. In this case to get better accuracy of the measuring system, it is preferable to implement the second method - to inject a known current and measure the voltage across the biological object.

To eliminate the negative effect of unknown contact impedance in bioimpedance measurement four-electrode system is used [4], where there are two pairs of electrodes: one pair, so called outer electrodes, are used for current injection and the second pair, called inner electrodes, are used for measurement of the voltage across the object. The bioimpedance measuring system can be divided on two main parts: the injecting part and measuring part. The injecting part consists of two functional units: the voltage waveform source (sine generator) and voltage controlled current source (VCCS).

Once the sine wave is produced by the voltage waveform source it must be convert into a current whose magnitude is unaffected by load (patient's bioimpedance). In a bioimpedance measuring system the function of voltage- to- current conversion is performed by the VCCS. An important parameter of a current source is its output impedance (Zout), which appears in parallel with the load (Z_L), connected to the current source output. Figure 1 shows a model of a current source.

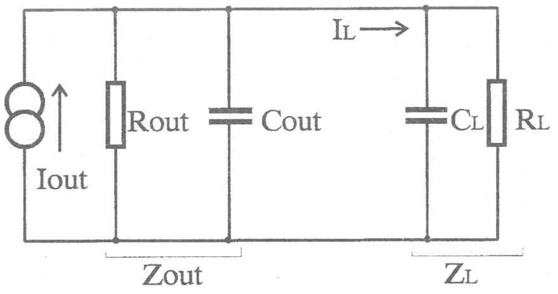

fig.1

For a perfect current source, output impedance (Zout) will be infinite and the load current (I_L) will not vary with changes in Z_L. For non-ideal (real) current source a finite output impedance causes an error to exist between the current that is sourced (Iout) and the current applied to the load (I_L). To minimize this error Zout must be sufficiently large. From a current source model, shown in figure 1, we can obtain the equation for I_L.

$$\frac{Zout.Z_L}{Zout + Z_L}.Iout = I_L.Z_L$$

$$I_L = \frac{Zout}{Zout + Z_L}.Iout \qquad (1)$$

We would like to find the minimal value of Zout, so that the load current changes in the limit of predetermined error over the range of possible load impedance. For example, if we use in the measuring part of the system 8(10) bits ADC, this means that the error , determined for 8(10) bits accuracy, will be $\varepsilon = 1/2^{8(10)}. 100\% = 0.25\%(0.1\%)$. Using equation (1) for two different load impedances Z_{L1} and Z_{L2}, we will get two different load currents I_{L1} and I_{L2}:

$$I_{L1} = \frac{Zout}{Zout + Z_{L1}}.Iout \qquad (2)$$

$$I_{L2} = \frac{Zout}{Zout + Z_{L2}}.Iout \qquad (3)$$

If we combine these two equations, the equation for the value of Zout could be obtained

$$Zout = \frac{I_{L2}.Z_{L2} - I_{L1}.Z_{L1}}{I_{L1} - I_{L2}} \qquad (4)$$

From this equation we can calculate the minimal value for Zout, which will meet our requirements about the accuracy of the measurement, by substituting the maximal and the minimal load for Z_{L2} and Z_{L1} respectively, desired load current for I_{L1} and the current that differs from the desired current by ε for I_{L2}. In this case for the range of bioimpedance changes from 0 to 2kΩ and for 8(10)bits accuracy, we will receive 512kΩ (2MΩ) for the minimal necessary value of Zout.

Additional requirements for implementation of the VCCS include:

direct-current blocking, temporal stability, ability to handle loads that include the capacitive components over the range of working frequencies. The capacitive components of the load in a real measuring system are: the series capacitances of the electrode/skin contacts; stray capacitances between electrodes; stray capacitances of the cables connecting the VCCS to the patient; etc. In a practical bioimpedance measuring system all these unwanted capacitances degrade the results from the ideal circumstances. An equivalent model of the real load for the VCCS in a bioimpedance measuring system, where the pointed components are considered, is shown in figure 2.

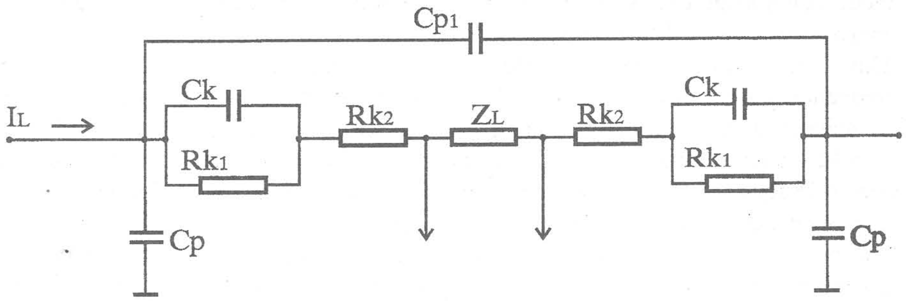

fig.2

where:

Z_L	- the impedance of the object (bioimpedance);
C_K, R_{K1}, R_{K2}	- represent the electrode/skin contact impedance;
C_{P1}	- stray capacitance between injecting electrodes;
C_P	- stray capacitance in the cables.

To reduce interference pickup the cables between the VCCS and the patient should be shielded. If grounded shields are used, the stray capacitance of the cable will act as a shunt capacitance for the output of the VCCS. The stray capacitance of the coaxial cables is about 100pF, which means that it will degrade significantly the performance of the high output impedance of the VCCS. If the shield is driven with approximately the same signal as the inner cable, a much smaller current will flow through the cable's capacitance. This technique reduces the effective cable capacitance to about 1pF. In practical implementation the shield may be driven by additional buffer (a voltage follower op-amp), which low output impedance preserves the advantage of the grounding.

Different techniques are used to implement the VCCS for bioimpedance measuring systems: VCCS using an inverting op-amp; transformer-coupled op-amp; Howland configuration op-amp; current-mirror based current source; etc.

The simplest version of a VCCS is an inverting operation amplifier with the load in the feedback path, as it is shown in figure 3. The output impedance is given by:

$$Zout = (1+G).Ri \qquad (5)$$

The value of Ri is limited by the input voltage (Vin) and the required output current (I_L). The open-loop gain (G) of the op-amp is a function of the frequency. The gain is very limited and the output impedance (Zout) decreases at higher measurement frequencies. The specific feature of this circuit is that the load must be floating, so it can not be connected to the reference ground and the contact impedance of the two current electrodes delivers a high common mode voltage to the input of the voltage measurement part of the system. To avoid high common mode voltage and to ensure the safety of the patient, a transformer-coupled current source can be used. This circuit is shown in figure 4.

This circuit can not provide high output impedance over a wide range of frequencies, because of the properties of the transformer, resulting from the stray capacitance and the limited transinductivity of the transformer.

Another implementation of VCCS, known as Howland type current source, is shown in figure 5. This circuit can be used as a base structure for implementation of a differential current source, as it is shown in [2].

The most important factor in this case is to keep the balance between the two current sources, connected in series. Any imbalance in the parameters of the two Howland type VCCS will result in a non-nominal value of the current injected through the patient and a common mode signal.

Another design of VCCS with high output impedance is based on op-amp supply current sensing (current-mirror) technique [1], [3]. The basic circuit is shown in figure 6. In this circuit the op-amp power supply leads are connected to the power supplies (+Ucc, -Ucc) via current mirrors. CM1 is a current mirror source which connects the op-amp's positive supply to +Ucc via the low input impedance current mirror source. CM2 is a current mirror which connects the op-amp's negative supply to -Ucc via the low input impedance current mirror. The resistor R, which is the load of the op-amp, is the current determining resistor. The op-amp is in a voltage follower configuration and it supplies a current I_R to the resistor R. This current is determined by the input voltage Vin according to the Ohm's low I_R= Vin/R . The current provided by the output stage of the op-amp has to be drawn from its supplies. It is equal to I_R and independent of the load Z_L. The advantages of this circuit are high stability, high output impedance and wide-band operation. The drawbacks arise from the non-idealities of the current-mirrors and the amplifier.

After this review of some practical implementations of voltage controlled current sources, used in bioimpedance measuring systems, we can conclude that it is difficult to achieve in practice high current accuracy over the range of possible bioimpedance changes, especially for the higher frequency, where the stray capacitances degrade significantly the performance of the current source.

An alternative approach, which can be implemented, is to use a less ideal current source and to measure the injected current. This approach is particularly attractive as the requirements about the quality of the current source are not so strong.

The classical method for current measurement is to introduce a known value high quality resistor in series in the current path and measure the voltage across the resistor. The modification of the load circuit, connected to the output of the current source, is shown in figure 7. In this case we have to take into consideration the following two requirements: First that for an accurate differential voltage measurement a differential amplifier with sufficiently high common mode rejection ratio (CMRR) is necessary. Second that the input capacitance of the instrumentation amplifier, which also influence (degrade) the output impedance of the current source has to be as small as possible.

REFERENCES

1. Denyer C.W. et al., A High Output Impedance Current Source, Physiological Measurement, 15, pp.79-82, 1994.

2. Jossinet J., Tourtel G., Jarry R., Active Current Electrodes for in vivo EIT, Physiological Measurement, 15, pp.83- 90,1994.

3. Lidgey J. et al., Developments in Current Driver Circuitry, Conference on Electrical Impedance Tomography, pp.183-189, Copenhagen, July, 1990.

4. Webster J.G.(editor) et al., Electrical Impedance Tomography, Adam Hilder, 1990.

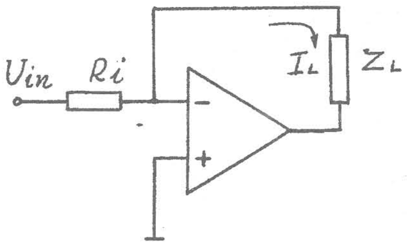

FIG. 3

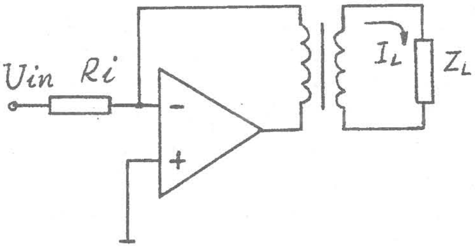

FIG. 4

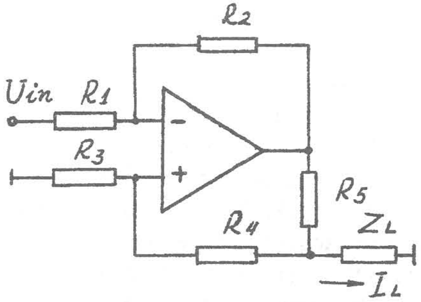

FIG. 5

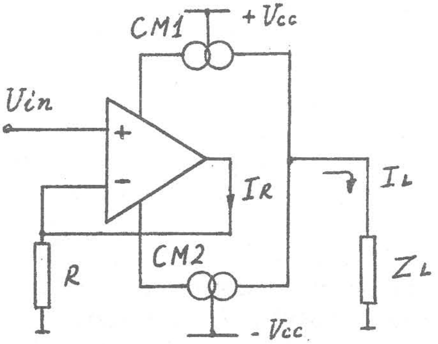

FIG. 6

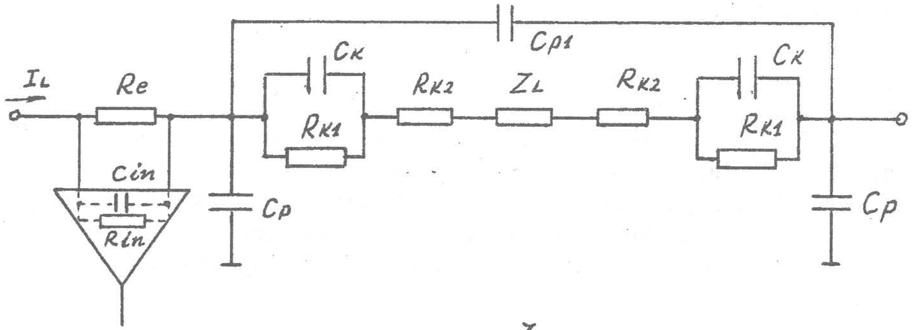

FIG. 7