Electrode Circuits for Frequency- and Code-Division Multiplexed Impedance Tomography

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Abstract — Traditional impedance tomography measurement systems make sequential four-terminal impedance measurements. Potentially faster frequency- and code-division multiplexed impedance measurement systems require that simultaneous current injection and voltage measurement take place at all terminals, making four-terminal measurements difficult. We describe an electrode interface circuit that simultaneously implements balanced current injection, and current and potential measurement, allowing four-terminal measurements on sets of multiple electrodes using these faster techniques. Circuit results show that accurate simultaneous four-terminal measurements are possible, thereby enabling faster impedance tomography systems.

Index Terms— electrical impedance tomography, bioimpedance, electrical impedance spectroscopy, resistometry.

I. INTRODUCTION

ELECTRICAL impedance tomography (EIT) and spectroscopy (EITS) are methods of imaging which use the different complex impedances of regions of the imaged object as the contrast variable [1,2]. The word tomography refers to the common method of imaging the object in planes (or, if the object is liquid, passing it in a pipe through the image plane). The planes are defined by the geometrical arrangement of the electrodes which connect the impedance measuring circuit to the imaged object. In tomography (EIT), the impedance at a single frequency is used, whereas in spectroscopy (EITS) the frequency dependence of the impedance is also used to enhance contrast. EIT is most commonly used in industrial applications for quantifying flow constituents and flow behavior, and EITS in medical applications for non-invasive imaging within the body.

For a typical EIT system, a ring of 8, 16 or 32 electrodes is used. Current is injected through pairs or electrodes and the potentials induced by the currents are measured at other pairs of electrodes. There are many different possible injection and measurement permutations. In perhaps the most popular method, current is injected through adjacent pairs of electrodes and the potential measured in all permutations of all the other electrodes (although, because each adjacent pair is characterized by their use for current injection, they may be omitted in the potential measurements to avoid redundancy).

A problem of EIT is that the resolution depends strongly on the number of measurements made; so a large number of linearly independent voltage measurements are required. For an 8-electrode system using the most minimal adjacent-pair sampling, 28 measurements are performed; and this number increases with the square of the number of electrodes. The problem is exacerbated in EITS, where measurements must be made at a number of different frequencies. Given that the subjects imaged (e.g. turbulent flows, or human thoracic cavities) are seldom static, a high frame rate or rate of image acquisition is desirable.

Three approaches have been used to increase the frame rate. The first is to optimize the sequential measurement process, which is time-division multiplexed or TDM EIT; Wilkinson and colleagues have achieved frame rates of 1000/s this way [3]. The second method is to separate the measurements in the frequency domain - so-called frequency-division multiplexed or FDM EIT [4–6]. This involves the simultaneous injection of currents at different frequencies, and demodulation of the induced voltages to derive separate measurements. A third method, recently proposed by the present authors, is to use orthogonal pseudorandom codes to separate the measurement channels – code division multiplexed or CDM EITS [7]. This involves modulating simultaneously injected currents with pseudorandom codes, and demodulating the induced voltages using the same codes. This method lends itself to spectroscopy as the frequency response of the system is directly accessible in the demodulated signals. The FDM and CDM methods have not yet received widespread adoption but we anticipate that the demand for increased frame rate will lead in this direction.

A problem faced in the FDM and CDM methods, and in the wider field of impedance measurement, is that accurate impedance measurement requires a four-terminal approach, which traditionally requires the separation of current-injection and potential-measurement electrode pairs. If potential measurements are made on electrodes whilst current is being injected through them, the potential measured includes the voltage drop across the contact resistances, which is likely to be significant in most EIT applications. This is shown schematically in Fig 1.

If the same electrodes are to be used simultaneously for current injection and voltage measurement in FDM and CDM systems, then the four-terminal method becomes difficult. One alternative is to use separate sets of electrodes for current injection and voltage measurement. This method was adopted in the FDM systems to date, as well as in some TDM systems [8]. Unfortunately, this doubles the number of electrodes required for a given number of measurements. The reduced electrode size reduces the accuracy of the impedance.
measurement and increases the practical difficulties in wiring, mounting and isolating the electrodes.

In this paper we present a method and circuit whereby electrodes can simultaneously be used for current injection and voltage measurement, in CDM and FDM systems, while correctly implementing the four-terminal principle.

Fig. 1: Comparison of two-terminal (A) and four-terminal (B) measurements. In A, the contact impedances \( Z_{c1} \) and \( Z_{c2} \) are inseparable from the sample impedance \( Z_S \), so \( V = I_S (Z_{c1} + Z_{c2} + Z_S) \), whereas in B, the high input impedance of the voltmeter renders negligible the current through the contact resistances \( Z_{c3} \) and \( Z_{c4} \) so that \( V = I_S Z_S \). \( Z_u \) are unknown bulk impedances.

II. METHOD

A circuit for an electrode interface of this kind must have the following characteristics:

- It must allow a defined or measurable current to be injected pair-wise with another electrode.
- It must allow a voltage, induced by a signal orthogonal to the injected current, to be measured without including the voltage drop across the electrode contact resistance.
- All electrodes must be included in any sequence such as an adjacent-pair measurement strategy.
- It must be able to operate within reasonable current, voltage and bandwidth levels, as defined in terms of the sample impedance and the desired frame rate.
- It must be compatible with the principles of safety in biomedical instrumentation.

These characteristics are examined below.

A. Current Injection

Given any two orthogonal signals \( f(t) \) and \( g(t) \), it appears trivial to inject currents proportional to each and measure the induced voltages at some other point. The situation is complicated by the need for all terminals to act simultaneously in current injection, as illustrated in Figure 2.

Fig. 2: This illustrates the problem of sharing electrodes amongst multiple current injection loops, each injecting orthogonal currents \( i_n \). The voltage \( v_n \) and impedance looking into each terminal (at the circuit side of the contact impedance \( Z_{c3} \)) is a complex function of the injected currents all around the outside loop.

It can be seen in Figure 2 that the current sources are effectively connected in a ring. This places very stringent requirements on the sources; they must be “stiff” in the face of wide and unpredictable voltage and impedance fluctuations at their terminals; and given the complexity of orthogonal signals, they must be able to source and sink symmetrical (and bipolar) currents at high bandwidth.

In practice, even with separate current injection and voltage measurement terminals, it has been found that current sources are seldom sufficiently stiff to render current measurement unnecessary [8]. This adds a complication to our specification, which is that we must be able to measure the currents as actually injected.

B. Typical Specifications

The EITS applications envisaged for this method are those where the imaged object changes rapidly; for example, the circulating constituents inside a hydrocyclonic separator, or blood flow in a head or body cavity. The specification in terms of currents, voltages and bandwidths can be determined by considering the imaging problem. In the present example, we require a 16-electrode EITS system to obtain 10 frames / second with a spectral bandwidth of 10Hz - 5MHz, in a medical (human) imaging application. Our intention is to measure impedance changes in ischemic tissue which occur at low frequencies (in the order of 100Hz or less), and simultaneously measure changes occurring due to blood flow which can be detected at high frequencies (of the order of 1MHz or greater). The allowable current levels are set by medical safety standards, to about 1mA. The combined impedance of the contact and tissue represent a load of up to 2k\( \Omega \), leading to a voltage compliance range of 2V for the source. The sample impedance ranges from 1\( \Omega \)-100\( \Omega \), so a gain stage is desired to amplify small standing voltages of 1mV. It is crucial that this voltage is recorded with a high accuracy as we would like to measure changes in impedance.
of the order of 0.1% of the sample impedance.

III. THE USE OF PULSE TRANSFORMERS

We have chosen to use high-frequency pulse transformers as current sources in our solution. The advantages of transformers are that they source and sink symmetrical and bipolar currents without difficulty; and that they inherently provide galvanic isolation. By putting a small resistor in series with each transformer, we obtain a compact circuit in which the currents and induced voltages at 16 terminals can be measured using 32 single-ended ADC channels. The circuit is shown in Figure 3.

![Fig. 3: The arrangement of current sources, transformers, ballast resistors \( R \), and voltage measurement points \( v_m \), for four of the sixteen electrodes.](image)

The operation of the system can be described in terms of the elements shown in Figure 3. Consider the simultaneous measurement of the voltage \( v_{3,4} \) induced between terminals \( T_3 \) and \( T_4 \) by the current \( i_2 \) injected between terminals \( T_1 \) and \( T_2 \); and the measurement of the voltage \( v_{1,2} \) induced between terminals \( T_1 \) and \( T_2 \) by the current \( i_4 \) injected between terminals \( T_3 \) and \( T_4 \). If we assume that \( i_2 \) is modulated by \( f(t) \) and \( i_4 \) is modulated by \( g(t) \), where \( f(t) \) and \( g(t) \) are orthogonal or nearly so, so that for example in FDM \( i_2 = I f(t) \) and \( i_4 = I g(t) \) where \( f(t) \) and \( g(t) \) would be non-identical sinusoidal carrier waves and \( I \) is a constant; and if we indicate the voltage measured at terminals \( k \) and \( l \) by the current injected by transformer \( m \) as \( v_{k,l} \), then:

\[
v_{1,2} \mid i_2 = \left[ (v_1 - v_2) \times (v_{T4} - v_{T3}) / R \right]_{LPF}
\]

\[
v_{3,4} \mid i_2 = \left[ (v_3 - v_4) \times (v_{T2} - v_{T1}) / R \right]_{LPF}
\]

where \( \left[ \right]_{LPF} \) represents the low pass filtering of the expression in parentheses (this would be the standard synchronous demodulation algorithm for frequency-multiplexed signals). Other modulation / demodulation schemes can similarly use \( (v_{Tm} - v_0) / R \), where these are the two voltages measured either side of transformer \( m \)'s series resistor, as a representation of the current injected for the purposes of demodulation.

IV. PROOF OF CONCEPT

The circuit shown in Figure 4 was used for a series of proof-of-concept tests.

![Fig. 4: The circuit used for proof-of-concept tests. The signal source for \( G_1 \) and \( G_2 \) (the pseudorandom codes) was a simple microcontroller. The network formed by \( R_1-R_5 \) and \( C_1 \) represents the loading impedance network of the imaged object. The transformers were ultrahigh bandwidth pulse transformers and the op-amps were high slew rate, low noise, FET input amplifiers.](image)

Two transformer coupled sources were tested on an RC load network. A microcontroller was used to repeatedly generate the two near-orthogonal 1023 bit length Gold codes from a look-up table. The terminal voltages were recorded using a digital storage oscilloscope (Figure 6). An impulse response was calculated for each channel using cross-correlation and the Fourier transform provided the spectrum of the impedance for each channel (Figure 7).

![Fig. 5: The injected currents generated by two transformers coupled as shown in Fig. 4. The currents were calculated by measuring the voltage across a series resistor (for example, measuring \( V_{T2} - V_4 \) in Fig. 4). The bit structure of the codes can clearly be seen; nonetheless, the sources are insufficiently stiff to allow nominal values to be used for demodulation purposes, and must actually be measured.](image)

![Fig. 6: Currents from the coupled circuits in Fig. 4, whereas Figures 8 and 9 show signals from the two channels decoupled. This allows us to measure the success of demodulating supposedly orthogonal channels which are carried simultaneously in the network.](image)
V. DISCUSSION

We have demonstrated that two coded, orthogonal channels may simultaneously measure impedance, provided the electrode interface circuits are designed to allow it.

REFERENCES