Active Electrode Based Electrical Impedance Tomography System

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Abstract: EIT can image the distribution of ventilated lung tissue, and is thus a promising technology to help monitor patient breathing to help selection of mechanical ventilation parameters. Two key difficulties in EIT instrumentation make such monitoring difficult: 1) EIT data quality depends on good electrode contact and is sensitive to changes in contact quality, and 2) EIT electrodes are difficult and time consuming to place on patients. This paper presents the design and initial tests of an active electrode based system to address these difficulties. An electrode belt is designed incorporating 32 active electrodes, each of which contains the electronic amplifiers, switches and associated logic. Tests show stable device performance with a conventient ease of use and good imaging ability in volunteer tests.

1 Introduction

Electrical Impedance Tomography (EIT) determines images of the conductivity within a body from electrical stimulation and measurement at electrodes placed on the body surface. EIT has shown significant promise as a technology to monitor the distribution of air within the lungs of mechanically ventilated patients. Such patients can have highly heterogeneous lungs, and data which EIT can provide are helpful to select lung protective ventilator settings. EIT fills a unique niche, by providing tomographic information without a bulky system or ionizing radiation.

Although EIT shows dramatic promise, it is still largely a technology in the research phase. We identify two key factors which impact the reliability and convenience of use of EIT technology. First, EIT is very sensitive to issues associated with the patient interface (electrodes and cables). The electrode contact with the body tends to have relatively high contact impedance, Z_c, which also varies with time due to body movement and drying of the contact. Such high Z_c means that all interference sources have a large effect; these include common mode gain, thermal noise, and EM interference and crosstalk between cables. Second, EIT is often inconvenient to apply since many systems require individual placement of electrodes – which can be especially difficult on the back of heavy or high-risk patients.

This paper describes design and initial testing of a system to address these factors. System design is based on an electrode belt containing active electrodes with minaturized supporting electronics mounted immediately onto the electrodes. The active electrodes system is motivated by the work of Rigaud *et al* [1], Li *et al* [2] and Guardo *et al* [3].

2 Hardware design

The design concept was to put the electronics as closely as possible to the patient to minimize problems related to the analog transmission of signal on high impedance lines.

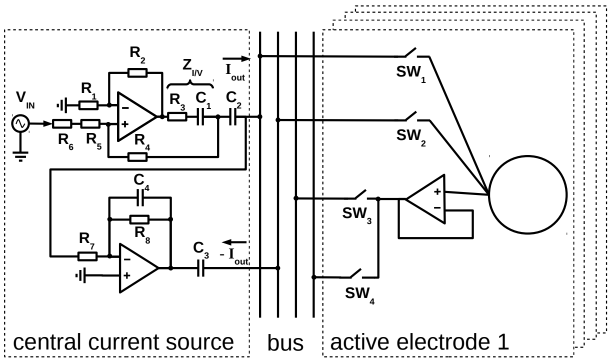

Figure 1: The central current source with the feedback circuit (on the left side of the bus) and the basic elements of an active electrode (on the right side of the bus) are depicted. Capacitance $C_{1-3} = 220\ nF$ are the DC blocking filter. Resistors $R_{3,6} = 390\ \Omega$ determine the conversion factor between voltage and current. Resitors $R_{1,2,5,4} = 10\ k\Omega$. Resistors $R_{7,8} = 5.6\ k\Omega$ are chosen equal to get a unity gain inverter. Capacitance $C_4 = 15\ pF$ avoids the oscillation of the circuit at high frequency.

The most direct way to accomplish that is to place an Analog to Digital Converter (ADC) and Digital to Analog Converter (DAC) on each electrode and communicate in the digital domain with the rest of the system. Unfortunately, it was not possible to meet the price and size requirements with multiple DACs Therefore we decided to move signal generation and measurement to a central Sensor Belt Connector (SBC). Since the analog signal should not be transmitted from the electrode to the SBC on high impedance lines using simple lead wires, the solution was to implement a voltage buffer as close as possible to the electrode.

The selected system architecture is shown in Fig. 1. Active electrodes are fabricated into a supported belt which surrounds the thorax, which allows placement of amplifiers and circuit logic as close as possible to the body. Having a voltage buffer close to the electrode has two beneficial effects: 1) it increases the input impedance of the electrode and 2) it reduces the length of high impedance paths. The input inpedance of an active electrode is determined by the stray capacitance of the buffer (2 pF) and other active elements (total about 15 pF). The input impedance of the same electrode but without active elements with 1 m bus cable was measured to be at least 200 pF (i.e. stray capacitance of the cable with adjacent ground lines). Being able to transmit the signal on low impedance line also reduces the coupling of external signal on the bus line and the cross-talk between the bus lines. The acquisition chain is shown in Fig. 2. In practice an electrode belt will tend to get dirty, for example sweat or blood could enter in contact with the belt structure or the electronic and induce damage. Thus, there is an hygiene requirement for the belt to be cleaned (sterilized) between patients. Since washable/sterilizable electronics are still experimental and their reliability not yet demonstrated, we recommend a one-time use belt strategy.

The electrodes: Each active electrode is referred to as a "node". Nodes could, in principle, drive multiple active electrodes, but our design uses one node per electrode. The tasks of an active electrode are: 1) injecting a current, 2) sinking a current, 3) buffering a voltage and multiplex the measurement on the analog lines of the bus using (SW_3 and SW_4) and 4) do nothing. Each node is managed by a embeded micro-controller (Atmel Atmega328) which contains the state table of the node. At start-up each node takes its reset value and waits for a pulse signal broadcast by the central unit on the sync line.

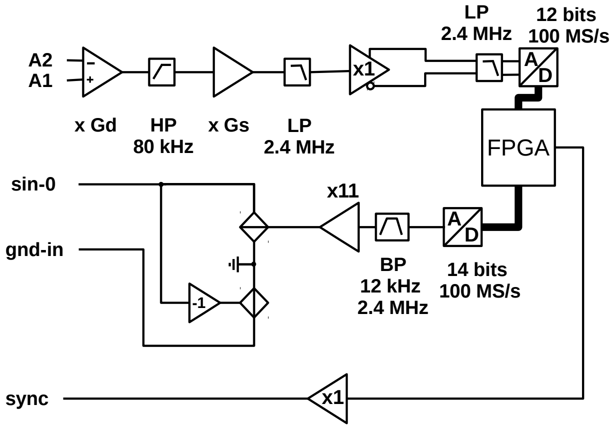

Figure 2: The implementation of the acquisition chain and signal generator on the SBC. The gain of the instrumentation amplifier, G_d, is set in combination with the gain, G_s, of the second amplifier stage to best fit the voltage input range of the ADC.

Each time the node measures a pulse, it triggers an interrupt, which fetches the next state in the state table and applies it to the analog switches.

The central current source: The voltage signal reference is generated using a DAC and is subsequently buffered to be supplied to a transimpedance circuit, the current source [4] itself. After flowing through the body the current needs to be sinked. The first approach consists of simply sinking the current to ground. Tests showed this approach gave high common mode voltage errors, so a third circuit, a commanded voltage source, is used to "actively" sink the current and balance the injection voltage, and significantly reduce the common mode voltage. This circuit has advantages over symmetrical current sources which injecting current with 180^o phase difference. These systems face the challenge to exactly match source and sink current levels, and typically require an additional external ground electrode to absorb current differences. For electrical safety, decoupling capacitors are placed in the current source circuit to avoid the undesired injection of direct current into the body.

3 Volunteer Tests

The designed system has been tested in terms of its electronics performance and behaviour in phantom tests. We have also performed numerous tests of human volunteers. Results show that the system is quick to place onto the subject, and normally functions immediately with very few poor electrode contacts. In some cases, a few electrodes are identified that give poor signals; this is addressed by pressing on the electrodes or by applying skin preparation gel. The time to place the belt and begin data acquisition is consistently under 1 min. We present an example of images acquired on a male healthy volunteer. Data were obtained at a frame rate of 15 frames/s and images were reconstructed using a linear time difference Gauss-Newton reconstruction algorithm using EIDORS [5]. Fig. 3 illustrates an image sequence and the conductivity time curves at a sample heart and lung pixel. The "lung pixel" shows a decrease in the conductivity with the inhalation, whereas the "heart pixel" shows an increase in the conductivity. This effect is physiologically reasonable as spontaneous breathing decreases thoracic pressure resulting in increased cardiac filling. In a mechanically ventilated patient, on the other hand, "heart pixels" follow the lungs

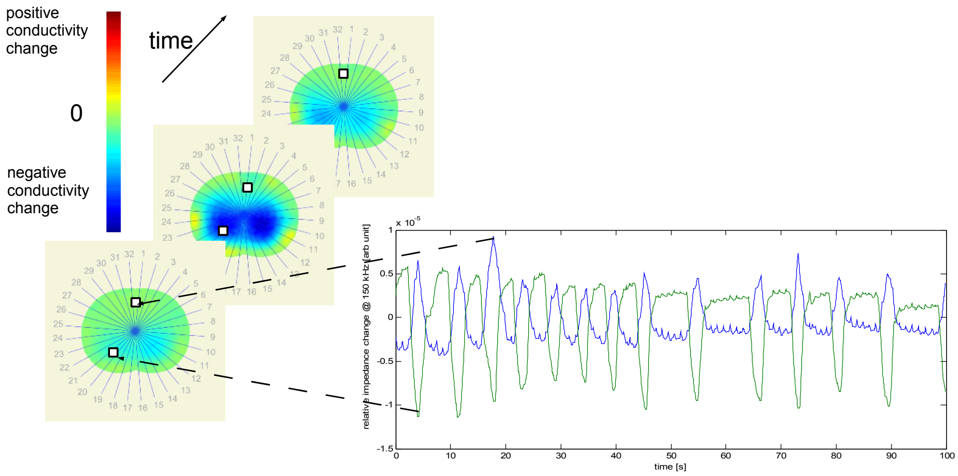

Figure 3: Temporal evolution of the EIT reconstructed impedance changes for two pixels one in the heart (blue) and one in the right lung (green) region.

because of pressure increases from the ventilator. Another effect is the movement of the heart during breathing which also affects EIT images.

4 Discussion

We present the design and initial tests of an active electrode EIT system, designed to address two key difficulties in EIT instrumentation: 1) EIT data quality depends on good electrode contact and is sensitive to changes in contact quality, and 2) EIT electrodes are difficult and time consuming to place on patients. An electrode belt was designed incorporating 32 active electrodes, each of which contains the required electronic amplifiers, switches and associated logic. Tests show stable device performance with a convenient ease of application and good imaging ability in volunteer tests.

5 Bibliography

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