PULSE MEASUREMENT PERFORMANCE IN DIFFERENTIAL RESISTIVITY IMAGE RECONSTRUCTION

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Abstract: Instead of sinusoidal current pattern for determine the resistive component of impedance, short current pulses are used and voltage responses are measured. An experimental tomography system with 16 electrodes is developed. Measurements are obtained in cylindrical saline-filed tank in order to achieve pulse measurements performance, i.e. spatial resolution, conductivity contrast and sensitivity. Duty-cycle of applied current pulses was lower then 0.01. Because of relatively high voltage amplitudes (several volts), the signal to noise ratio is very high and the high input amplifiers gain is not demanded. The whole measurement system is much simpler then in the case of sinusoidal current measurement. Problems caused by different and high skin resistance have been solved. Image reconstruction is done using back projection method along straight lines (inverse Radon transform) and an algorithm implementation for reducing data artefacts due to the non-straight current distribution.

Introduction

The aim of Electrical impedance tomography is to obtain image reconstruction of resisivity distribution across body cross section. Through ring of 16 or more electrodes, placed on the boundary of the object, current pulses are applied and voltage responses are measured. A major concern in this process is to choice appropriate currents to apply. Mostly sinusoidal current is applied and voltage are measured on non-current-carrying electrodes [1]. Measuring on current-carrying electrodes is usualy avoided to minimize the influence of skin resistance and capacitance.

We have chosen short current pulses with low duty-cycle as current pattern. Because of short current pulses and high skin capacitance, skin resistance can be considered as short-circuited and it does not affect the internal body resistance measurements. In this case voltage can be measured in current-carrying electrodes. Pulse duration is 5µs maximum and duty-cycle is lower then 0.01. Therefore, high current magnitude can be used (up to 30 mA) without any harmful influence on the tissue or muscle contractions risk. Despite of small effective current applied on tissue, measured voltage amplitudes are high enough to obtain very low signal-to-noise ratio.

Materials and methods

EIT measurement are made using ring of equally spaced electrodes placed on the boundary of the object. In this case measured impedance between two electrodes can be presented with well-known fifth elements scheme (Fig. 1.a) where R_p and C_p are resistance and capacitance of electrode-skin connection and R_s is internal body resistance. In most cases both electrode are the same and fifth element scheme can be reduced on three elements (Fig. 1.b) [2].

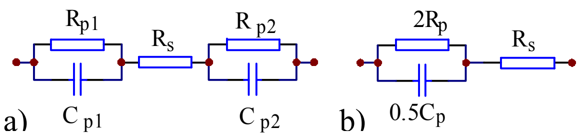

Fig. 1. Impedance equivalent circuits when R_{p1}=R_{p2}=R_p and C_{p1}=C_{p2}=C_p

The value of model elements can be obtained from voltage response on short current pulses [3]. In Fig. 2. a voltage response on current pulse of equivalent circuit with description are shown.

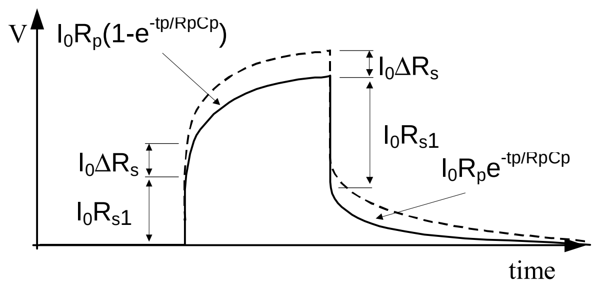

Fig. 2. Voltage pulse response of equivalent circuit

Pulse amplitude is I_0 and duration t_p. The rising edge in voltage response corresponds the internal resistance R_s, and if current pulse amplitude is known internal resistance can be calculated. Second, exponential part is a result of charging capacitance C_p with resistance R_p in parallel. The third voltage drop is a result of internal serial resistance R_s, as first part of voltage response I_0R_{s1}, but this drop is bigger because of chemical reactions occurred during charging period t_p.

The goal in resistive tomography is measurement of the internal resistance R_s for reconstruction of internal resistivity distribution. If internal resistance R_s has been changed, the change can be measured as the rising edge voltage step before charging of electrode capacitance C_p occurred. Comparison between measured pulse response and calculated from impedance measured with RLC meter shows deviation smaller than 8%, which can be neglected.

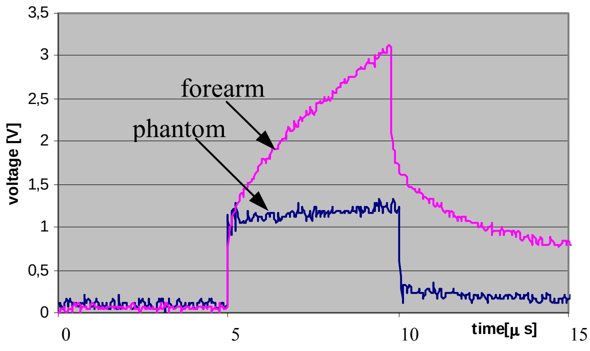

Fig. 3. Voltage response measured on forearm and phantom

Described equivalent circuit and theoretical voltage response gives very good congruity with waveshapes recorded on biological tissue [3]. In Fig. 3. voltage response measured on diametrically opposed electrodes on forearm and on saline filled EIT phantom are shown. In the second case there is no exponential voltage rising, because saline is in direct contact with electrodes and capacitance C_p can be assumed as very high and parallel resistance R_p very low. In case of measuring on the tissue, voltage must be measured near rising edge to avoid influence of capacitance or resistance of skin.

Experimental EIT hardware system

EIT measurement system [4] with 16 electrodes is based on ISA bus throw PC card. The whole system is programmable and PC controlled. Each electrode can be addressed as current source or current destination. Current generator is switched with analog switches to 16 electrodes. Only one electrode can act as the current source at the time, but more electrodes can act as the current destination if is necessary. Pulse amplitude can be selected between 1,2, 5 or 10 mA. Frequency and pulse duration are also programmable, but it depends of PC master clock. For 100MHz clock frequency range from 1,5kHz to 1MHz can be obtained with minimal pulse duration of 1μs.

For voltage response measuring electrodes are multiplexed to differential amplifier. Each electrode can be addressed as positive or negative. For avoiding constant voltage offset that contains no useful measured data, as a result of constant resistance, AD-DA converters are used. Digital value of measured voltage offset is saved, converted to analogue value and subtracted to other measured voltages to obtain only useful values. In this way differential data voltage resolution is greatly improved.

Special attention was paid on S&H circuit. Considering that amplitude of rising edge of voltage response is measured, S&H circuits should be as fast as possible. Used circuits have sampling time about 200ns.

Measuring and reconstruction algorithm

Measurements are obtained in a tank with 30cm diameter and 16 electrodes equally spaced on boundary of the tank. Tank is filed with distilled water with different volume of NaCl to obtain different conductivity. Data collection method [1] is the opposite method with some modifications. Bipolar measurements are used what means that same electrode pair addressed to a current source is used for voltage measurement. Instead of using only diametrically opposed electrodes and rotating it, all opposed electrode with parallel current path (1-16; 2-14;…7-9) are measured as one set of data collected at angle φ_n.

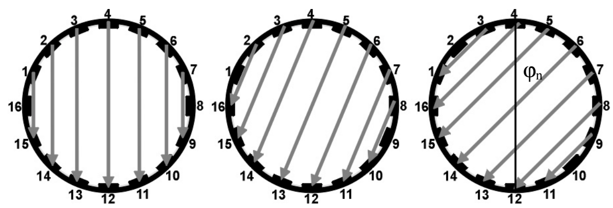

Fig. 5. Measuring algorithm (three steps)

The next set of data is collected by clockwise rotating measurement pattern to next electrode pair (2-16; 3-15;…8-10) as is shown in Fig. 5. For 16 electrodes, after rotating for 360°, 112 measured data can be obtained even some of them are redundant.

Seven measured voltages at projection angle φ_n are stored as projection vector $\boldsymbol{p}_{\varphi n}$. For 16 projection angles data matrix is defined as $\boldsymbol{D}=[\boldsymbol{p}_{\varphi 1}\ \boldsymbol{p}_{\varphi 2}\ldots\ \boldsymbol{p}_{\varphi 16}]$. For differential resistivity reconstruction two measured data are obtained. The first one $\boldsymbol{D}_H$ on homogenous medium (only water) and the second one with inhomogeneous object placed into the tank, denoted as $\boldsymbol{D}_I$. Differential data are calculated as $\boldsymbol{D}_D=\boldsymbol{D}_I - \boldsymbol{D}_H$ and in general, back-projected to reconstruct resistivity distribution. Reconstruction has been made using MATLAB Image Processing Toolbox and algorithm is shown in Fig. 6. Before reconstruction, projection matrix $\boldsymbol{D}_D$ has only 7 measured points per projection angle and additional points must be interpolated to reach better image resolution. Projection data were interpolated to 50x16 matrix $\boldsymbol{D}_D^*$ using cubic spline interpolation.

In opposite to the x-ray Computed Tomography where x-rays have straight paths, current flow has not. For this reason measured data does not represent resistance only between specified electrodes but resistance of neighboring electrode pair is partially included. To reduce those artifacts interpolated projection vectors $\boldsymbol{D}_D^*=[\boldsymbol{p}_{\varphi 1}^*\ \boldsymbol{p}_{\varphi 2}^*\ldots\ \boldsymbol{p}_{\varphi 16}^*]$ are convoluted with Gaussian function. As a result, projection vectors $\boldsymbol{D}_{proj}$ have narrower inhomogeneity lattice with higher amplitude.

Filtered back projection [5, 6] through 16 measured angles was performed on column vectors in matrix $\boldsymbol{D_{proj}}$. Before back projection each column was filtered with Shepp-Logan filter. Image was reconstructed in 60x60 resolution.

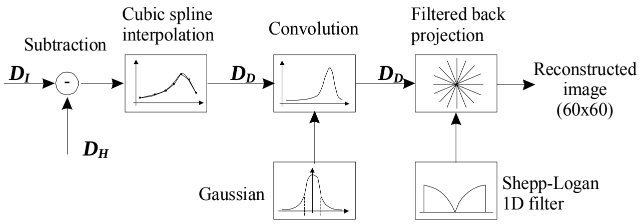

Fig. 6. Reconstruction algorithm

Results

For experimental measurement, tank was filled with water with resistivity of 420Ωcm. Piece of wet wood 2x3cm size, with resistivity about 15kΩcm was placed into tank at different places from boundary to tank center. After measurement using current pulses with amplitude of 2mA, pulse duration of 5µs and described reconstruction algorithm, resistivity distribution showed in Fig. 6. is obtained.

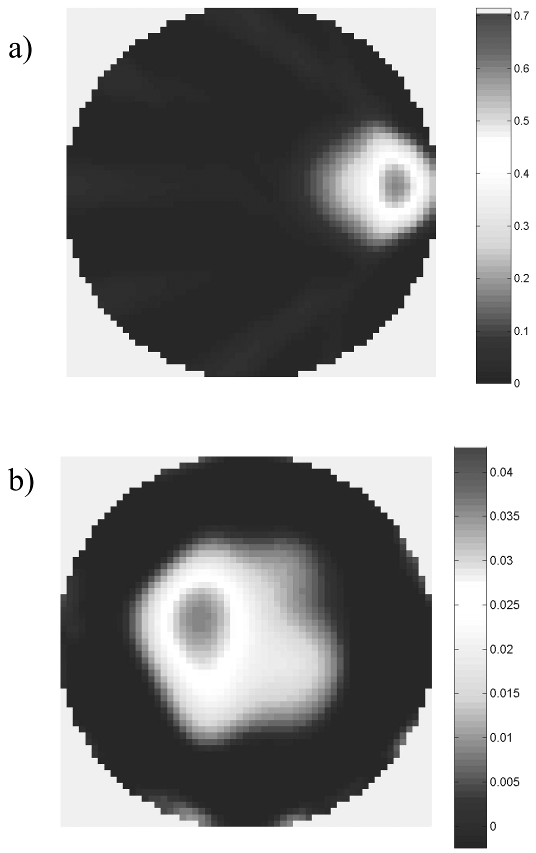

Fig. 6. Reconstructed image distribution with inhomogeneity in boundary region (a) and center (b) of the tank

The resistivity contrast is ratio between calculated homogenous water resistance and maximum value of calculated resistance. However, it has been changed in dependence where inhomogeneity is taking place. As the inhomogeneous object is closer to the electrodes, higher contrast can be obtained. Contrast dependence of the object distance from electrodes and current amplitude is shown in Fig. 7. It can be seen that resistivity contrast has been changed more then factor 10 and for distance higher than 6cm from boundary (20% of tank diameter) resistivity contrast is very small. Consequently reconstructed image is blurred and inhomogeneity takes much higher area. However, increase of current amplitude cause contrast increasing as well, but only in boundary region and it is not linearly proportional to current amplitude.

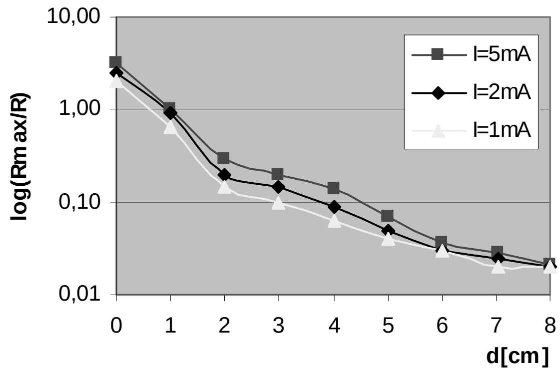

Fig. 7. Contrast dependence of inhomogeneity distance from boundary

The smallest detectable object in homogenous electrolyte is the isolator formed as a stick with diameter 0,5cm in boundary region and about 1cm in the center. The same results are obtained with object with smaller resistivity than electrolyte.

Pulse duration does not affect to sensitivity of measurement. For in vivo measurement it is important to reach duty-cycle as small as possible. In this case high current pulse amplitude can be used (better sensitivity) and tissue overloading is avoided.

Conclusion

Described measurement method using short current pulses can be used for measurement internal resistance in EIT. In this case current generator and whole system is much simpler than using sinusoidal current. High current pulse amplitude can be used as opposite to the amplitude of sinusoidal current in impedance tomography. As a result voltage response is high enough, measured signal is without noise, so high voltage gain of amplifiers is not required. Because of relatively high resolution during measuring and small signal to noise ratio, quite good resistivity image distribution can be obtained with simple filtered back projection method. Parallel opposite method for measurement gives better resolution and better contrast in boundary region. If inhomogeneity is expected in center region, different data acquisition method and image reconstruction should be used.

Acknowledgment

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