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Super-resolution reconstruction of electrical impedance tomography images☆

Ricardo Augusto Borsoi*,[a], Julio Cesar Ceballos Aya[b], Guilherme Holsbach Costa[b], José Carlos Moreira Bermudez[a]

[a] *Department of Electrical Engineering, Federal University of Santa Catarina, Florianópolis, Brazil*
[b] *University of Caxias do Sul, Caxias do Sul, Brazil*

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ABSTRACT

Electrical Impedance Tomography (EIT) systems are becoming popular because they present several advantages over competing systems. However, EIT leads to images with very low resolution. Moreover, the nonuniform sampling characteristic of EIT precludes the straightforward application of traditional image super-resolution techniques. In this work, we propose a resampling based Super-Resolution method for EIT image quality improvement. Results with both synthetic and *in vivo* data indicate that the proposed technique can lead to substantial improvements in EIT image resolution, making it more competitive with other technologies.

1. Introduction

Real-time imaging systems are particularly important in medicine as they provide useful means of monitoring the patient in intensive care or during medical procedures like intraoperative image guidance. However, most intraoperative imaging systems are expensive, offer radiation risks and do not operate in real-time [1]. Likewise, real-time monitoring can avoid lung overdistension and collapse in mechanical lung ventilation, reducing ventilator-associated pneumonia rates [2].

Electrical Impedance Tomography (EIT) is an imaging method that leads to low cost, highly portable, real-time and radiation-free imaging devices. These characteristics make it suitable for bedside diagnosis and intraoperative imaging. However, the lower image quality when compared to other tomography methods still keeps EIT from being widely applied in medical diagnosis and guidance.

The image resolution in tomography methods such as Computed Tomography (CT) and Positron Emission Tomography (PET) has significantly improved with the direct application of Super-Resolution Reconstruction (SRR) techniques. SRR consists basically in the reconstruction of a high resolution (HR) image by extracting non-redundant information from several low resolution (LR) images (usually acquired in the presence of motion) of the same object. SRR was originally a technique intended for overcoming physical limitations of optical imaging sensors. More recently, it has been successfully adapted for other imaging systems such as CT [3], PET [4], magnetic resonance [5] and optoacoustic tomography [6], leading to the improvement of the resulting image quality.

Although SRR algorithms developed for optical systems require some adaptations for super-resolving tomographic images originating from PET or CT, the principle of operation of these systems is quite similar to that of optical image acquisition systems. An EIT system, on the other hand, is based on different operation principles. It employs diffuse electrical currents instead of a coherent X-ray beam to generate images, which makes the image acquisition process inherently nonlinear. Besides, the use of traditional finite element (FE) techniques for solving the EIT inverse problem leads to an image represented by a nonuniform grid of conductivity

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* Corresponding author.
E-mail addresses: raborsoi@gmail.com (R.A. Borsoi), jccaya@ucs.br (J.C.C. Aya), holsbach@ieee.org (G.H. Costa), j.bermudez@ieee.org (J.C.M. Bermudez).

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values, as opposed to the uniformly sampled images acquired by sensor grids used in optical systems. These characteristics preclude the direct application of traditional SRR methods to EIT.

Recently, a new SRR method has been developed for the reconstruction of optical images generated by hypothetical nonuniform pixel arrays called Penrose Pixels [7]. Penrose tiling is a method of aperiodic tiling that uses different geometric forms at different orientations to fill a plane [8]. Rhombus Penrose tiling is a specific type of aperiodic tiling that employs only two types of rhombus of equal side length. To allow the representation of the nonuniform pixel grid using matrix-vector notation, it is proposed in [7] that the LR images be upsampled to a regular HR grid as soon as they are acquired. The upsampling operation is dependent on the sensor layout map. The resulting image is called Intermediate HR (IHR) image. After upsampling, image registration and SRR algorithms can be applied on the IHR grid, where the processes of downsampling to and upsampling from the nonuniform grid are incorporated to the reconstruction algorithm by means of an acquisition model. This method has been shown to be capable of achieving larger magnification factors than those obtained using regular pixel arrays, as there are no constraints concerning (sub)pixel displacements between the LR observations because the grid has no translational symmetry. Unfortunately, however, such sensor arrays are currently not being manufactured, and this promising technique remains largely without practical application.

In this paper, a new super-resolution strategy for EIT images is proposed. By employing the resampling strategy used in [7] to deal with the aperiodic tiling characteristic of Penrose pixels, we developed a method for super-resolving images generated with arbitrarily irregular tiling such as EIT images, which are composed of non-uniformly spaced pixels forming a finite element mesh (FEM). The EIT data acquisition system is modeled as in an optical SRR problem with nonuniform sampling, and the image reconstruction is performed independently from the EIT inverse problem. This allows the proposed methodology to be used with arbitrary EIT algorithms and FEMs, which includes commercial EIT systems already in operation. Quantitative and perceptual evaluations show substantial improvements in the resolution of the reconstructed images when compared to the original ones in LR. This observation remains consistent for different choices of algorithms used in the EIT inverse problem.

This paper is organized as follows. In Section 2 the basic principles of EIT and SRR are explained. In Sections 3 and 4, the EIT imaging system is modelled as a nonuniform optical imaging system. Results illustrating the super-resolution of EIT images are presented in Section 5, along with a discussion about its complexity and robustness. Finally, Section 6 concludes this paper.

2. Operation principle of EIT and SRR

2.1. The EIT working principle

In electrical impedance tomography, the electrical conductivity $\sigma(\mathbf{p})$ inside a domain Ω is estimated from voltage and current measurements $\{V(\mathbf{p}), I(\mathbf{p})\}$ at its boundary $\partial\Omega$, where $\mathbf{p}$ denotes the spatial position vector. The physical process relating the electrical conductivity of the domain with the measurements is described by the following partial differential equations [2]:

$$\begin{aligned} &div\,(\sigma(\mathbf{p})\nabla V(\mathbf{p})) = 0\,, \quad \mathbf{p}\in\Omega \\ &\sigma(\mathbf{p})\frac{\partial V(\mathbf{p})}{\partial \boldsymbol{\nu}} = j(\mathbf{p})\,, \qquad \mathbf{p}\in\partial\Omega \end{aligned} \tag{1}$$

where $div(\cdot)$ is the divergence operator, $\boldsymbol{\nu}$ is an outwardly-pointing unitary vector normal to $\partial\Omega$ and $j(\mathbf{p})$ is the current density on $\partial\Omega$, satisfying $\int_{\partial\Omega} j(\mathbf{p})d\mathbf{p} = 0$.

Since the voltage and current values are measured through a finite number of electrodes attached to the surface of the body, this discretization effect must be represented in some form [9]. One simple model for the influence of the electrodes is the *gap model*, where the boundary condition of (1) is described as:

$$\sigma(\mathbf{p})\frac{\partial V(\mathbf{p})}{\partial \boldsymbol{\nu}} = \begin{cases} I_l/A_l\,, & \mathbf{p}\in\partial\Omega_{E_l}\,, \quad l=1,\ldots,L \\ 0\,, & \mathbf{p}\notin\cup_{l=1}^{L}\partial\Omega_{E_l} \end{cases} \tag{2}$$

where L is the number of electrodes, and I_l and A_l are, respectively, the current and the area of the lth electrode, which is denoted by E_l. The region of the domain boundary $\partial\Omega$ occupied by the lth electrode is denoted by $\partial\Omega_{E_l}$, and the voltage $V(\mathbf{p})$ at the center of $\partial\Omega_{E_l}$ is denoted by V_l. The condition $\int_{\partial\Omega} j(\mathbf{p})d\mathbf{p} = 0$ is replaced with $\sum_{l=1}^{L} I_l = 0$, and the integral of $j(\mathbf{p})$ over $\partial\Omega_{E_l}$ equals I_l.

Besides (1) and (2), one must also consider the effect of the high conductivity of the electrodes in the measured voltages. This can be performed either by simple models like the *shunt model*, which considers V_l to be constant on $\partial\Omega_{E_l}$, or using more elaborate models which take the contact impedance between the electrode and the body into account [10]. Along with the choice of a ground level such as $\sum_{l=1}^{L} V_l = 0$, this set of considerations constitute the so-called *complete electrode model* [10].

With this model, one can fully describe the behavior of the voltages $\mathbf{V}_F = [V_1,\ldots,V_L]$ in the electrodes at the object/domain boundary $\partial\Omega_{E_l}$ given the conductivity of the medium $\sigma(\mathbf{p})$ and a set of injected currents $\mathbf{I}_F = [I_1,\ldots,I_L]$. This process is described in compact form by a nonlinear forward operator defined by $\mathbf{F}(\sigma(\mathbf{p}), \mathbf{I}_F) = \mathbf{V}_F$, and constitutes the *EIT Forward Problem* [2].

Given a set of $L-1$ linearly independent voltage and current measurements $\mathbf{I}_F^k = [I_1^k,\ldots,I_L^k]$ and $\mathbf{V}_F^k = [V_1^k,\ldots,V_L^k]$, $k=1,\ \ldots,L-1$, the goal of the *EIT inverse problem* is to find the approximate conductivity $\mathbf{y}_\Delta = \hat{\sigma}(\mathbf{p})$ such that the voltages predicted through the forward problem $\widehat{\mathbf{V}} = \mathbf{F}(\hat{\sigma}(\mathbf{p}), \mathbf{I}_F^1,\ldots,\mathbf{I}_F^{L-1})$[1] are as close to the actual measurements $\mathbf{V}_M = [\mathbf{V}_F^1,\ldots,\mathbf{V}_F^{L-1}]$ as possible [2].

[1] Note that the forward operator $\mathbf{F}(\sigma(\mathbf{p}), \mathbf{I}_F^1,\ \ldots,\mathbf{I}_F^k)$ describes an arbitrary number of sets of voltage measurements $\mathbf{V}_F^1,\ \ldots,\mathbf{V}_F^k$, depending on the number k of input current measurements $\mathbf{I}_F^1,\ \ldots,\mathbf{I}_F^k$.