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A method to remove the projection error in triple–energy radiography with contrast medium

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Elsevier use only: Received date here; revised date here; accepted date here

Abstract

The density-map reconstruction of a radiological contrast medium is affected by noise arising from the background lack of homogeneity (the so-called “projection error”) if images of the medium are collected starting from quasi-monochromatic X-ray beams. This noise, especially for a dual-energy reconstruction algorithm, becomes more significant than the statistical fluctuations of the photon transmitted flux, dramatically reducing the accuracy and the sensitivity of the reconstruction. In this work, we investigate the efficacy of the triple-energy technique, which is based on the simultaneous acquisition of three monochromatic images of the same target injected with contrast medium. A theoretical analysis allows to estimate the sensitivity and the accuracy of the reconstructed density-map compared with the dual energy one (i.e., the density map reconstructed acquiring only two monochromatic images). To validate the theory, a set of experimental measurements were performed: results show that triple-energy drastically reduces the projection errors (from 10 to 60 times smaller to dual-energy one), making it negligible with respect to the statistical noise.

Keywords: X-ray radiography; monochromatic beams; multi-energy tomography

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1. Introduction

The triple-energy radiography with contrast medium is one of the techniques employing monochromatic X-rays to produce radiodiagnostic images. Since 1970s, many authors have described the peculiarities of the use of monochromatic X-rays in medicine, defining the improvements with respect to the traditional radiology based on polychromatic X-ray beams. First, the possibility to reconstruct the density-map of a target material (*e.g.* a contrast medium) represents an appreciable advantage, removing selectively the signal of all other tissues. This possibility had been successfully applied in dual-energy K-edge subtraction angiography [1,2]. Further, the use of monochromatic X-rays allows to extract signal which is a *linear* function of the mass-thickness of the target material, extending the *imaging* concept. In fact, a monochromatic image is not only a picture of morphological features of a sample, but it is a *measure* of the density-map of the basis materials that constitute the sample. It would be thus correct to speak of *absorptiometric imaging*, as extension of the dual- and triple-photon energy techniques, as applied to the diagnosis of osteoporosis. Many authors have investigated theoretical and experimental limits of the absorptiometry [3,4]. We focused our attention on the reconstruction of the mass-thickness of a target material (Iodine), in presence of a background of a mixed tissue (soft, bone). The triple-energy technique has many applications in diagnostic radiology, in particular for its high signal to noise ratio, and its high sensitivity to low concentration of contrast medium. At the Department of Physics of the University of Bologna (Italy), we are developing a CT system with an X-ray quasi-monochromatic source for multi-energy small animal studies [5,6]. One of our goals is to get an *in vivo* imaging of the cancer growth and metastasis development in different tumor types on mice. In this paper we present a brief description of the triple-energy system and an investigation of the efficacy of the triple-energy technique. We also present some simulated and experimental results which show the advantages of triple-energy for removing the projection error.

2. Materials and methods

2.1. Theoretical approach

In order to determinate the mass thickness of a target material (for example, a contrast medium), it is fundamental to extrapolate signals produced by other tissues surrounding the target, namely representing the background. For monochromatic X-ray beams, the logarithmic intensity attenuation is described by the well-known Beer-Lambert law for a two-component system (target and background):

$$T = \ln(I_0/I) = \hat{\mu}_{mdc} L_{mdc} + \hat{\mu}_{bck} L_{bck} \qquad (1)$$

where $\hat{\mu}_{mdc}$ and $\hat{\mu}_{bck}$ indicate the mass-attenuation coefficients of the contrast medium and of the background, respectively; L_{mdc} and L_{bck} are the mass-thicknesses, product of density and thickness. If $\hat{\mu}_{mdc}$ and $\hat{\mu}_{bck}$ are precisely known, L_{mdc} and L_{bck} could be computed by solving a system of two equations, obtained by the measurement of the logarithmic attenuations *T(1)* and *T(2)* at two energies:

$$\vec{T} = A\vec{L}, \quad \vec{L} = A^{-1}\vec{T} \qquad (2)$$

$$\vec{L} = \begin{pmatrix} L_{mdc} \\ L_{bck} \end{pmatrix} \quad \vec{T} = \begin{pmatrix} T(1) \\ T(2) \end{pmatrix} \quad A = \begin{pmatrix} \hat{\mu}_{mdc}(1) & \hat{\mu}_{bck}(1) \\ \hat{\mu}_{mdc}(2) & \hat{\mu}_{bck}(2) \end{pmatrix} \qquad (3)$$

Unfortunately the background is often not homogeneous, and it is characterized by several changes in tissues composition and thickness. The *dual-energy* technique corresponds to the rough approximation $\hat{\mu}_{bck} \equiv \hat{\mu}_1$, where $\hat{\mu}_1$ is a reference material (e.g. water). This imprecise representation of the background mass-attenuation generates the *projection error* ΔL_{mdc}, defined as the difference between the estimated value of the target mass-thickness and the true one. By increasing the number of used energies to three, it is possible to improve the representation of $\hat{\mu}_{bck}$. Indeed, the *triple-energy* approximation takes the following approximation:

$$\hat{\mu}_{bck} \equiv g\hat{\mu}_1 + (1-g)\hat{\mu}_2 \qquad (4)$$

where $\hat{\mu}_1$ and $\hat{\mu}_2$ are the reference materials (e.g. water and aluminium), while *g* is a mixing parameter, which has to be determined solving the system of the following three equations:

$$\begin{pmatrix} T(1) \\ T(2) \\ T(3) \end{pmatrix} = \begin{pmatrix} \hat{\mu}_{mdc}(1) & \hat{\mu}_1(1)-\hat{\mu}_2(1) & \hat{\mu}_2(1) \\ \hat{\mu}_{mdc}(2) & \hat{\mu}_1(2)-\hat{\mu}_2(2) & \hat{\mu}_2(2) \\ \hat{\mu}_{mdc}(3) & \hat{\mu}_1(3)-\hat{\mu}_2(3) & \hat{\mu}_2(3) \end{pmatrix} \begin{pmatrix} L_{mdc} \\ gL_{bck} \\ L_{bck} \end{pmatrix} \quad (5)$$

This formalism can be extended in principle to N energies, by using *N-1* reference materials for representing the background:

$$\hat{\mu}_{bck} \equiv g_1\hat{\mu}_1 + g_2\hat{\mu}_2 + \ldots + g_{N-2}\hat{\mu}_{N-2} + (1 - \sum_{i=1}^{N-2} g_i)\hat{\mu}_{N-1} \quad (6)$$

By solving the system (6) for each point of the sample, the background variations can then be taken into account, and the signal of the target material can be correctly reconstructed. The statistical errors affect both the accuracy and the sensitivity of the technique. They put an inferior limit to the mass-thickness of target material detectable. In fact, also in absence of projection error, the mass-thickness of the target (to be detected significantly different from zero) must be larger than the statistical confidence interval assumed. We performed a first simulation with a phantom composed by overlapped layers of soft tissue (thickness = 5 cm, ρ = 1.06 g/cm^3), bone (two layers of thickness = 0.5 cm, ρ = 1.92 g/cm^3), iodine in water solution (thickness = 0.5 cm, ρ = 0.2 g/cm^3). Statistical fluctuations have been simulated according to Poisson statistics. We simulated the acquisition of three monochromatic images, at 30 keV, 40 keV, and 80 keV. The projections over the basis materials (Water, Aluminum and Iodine) have been computed. The mass attenuation coefficients used for the simulations have been taken from the NIST database. We computed the projection error, as a function of the Iodine concentration. The background has been simulated with four different compositions, varying the fraction of soft and bone content.

2.2. Experimental set-up

To validate the theoretical model for triple-energy, we carried out two distinct set of measurements:

- Fixed Iodine concentration: in the first set, we used a sample constituted of a plastic cuvette containing a water solution of Iodine with concentration 100 mg Iodine/cm^3. The thickness of the solution crossed by the beam was 10 mm. Then, we added to the sample, sequentially, layers of Aluminum of increasing thickness. For each thickness, we measured the set of logarithmic attenuation values at 30 keV, 40 keV and 50 keV. These values have been used to reconstruct the mass-thickness of Iodine with triple-energy matrix. The first and the second energies have been used to compute dual-energy reconstruction.

- Fixed background composition: in the second set of measurements, the sample was constituted of a plastic cuvette with a 2 mm-thick Aluminum plate on the front (fractional mass-thickness about 35% Al, 65% water). We measured the logarithmic attenuation of the beam at 30 keV, 40 keV and 50 keV, for many different concentrations of Iodine (from 0.77 mg/cm^3 to 100 mg/cm^3). Reconstructed mass-thicknesses of Iodine were computed for dual-energy and triple-energy.

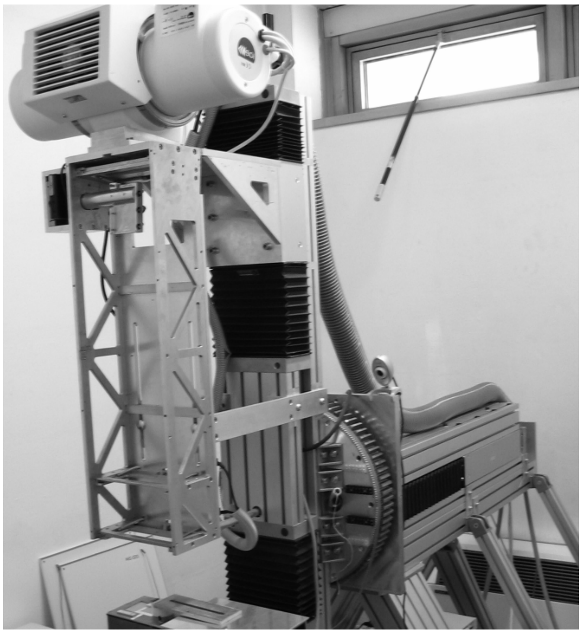

Figure 1: Picture of the multi-energy CT system: the quasi monochromatic beams are obtained by means of the Bragg monochromator coupled to the X-ray tube.

The experimental facility used for measurements consists of a tungsten anode tube (Gilardoni CHS) with high precision and stability, coupled to a Bragg monochromator based on a LiF(220) crystal. The beam, emerging from the crystal, passes through a set of collimators and is directed towards the sample. The movements of the crystal, for the selection of the energy, is driven by a motorized goniometric system, and controlled via software. The detector used for measurements is an XR-100T-CdTe (Amptek), coupled to a multichannel analyzer MCA-8000A (Amptek). In Figure 1 a picture of the experimental system is shown.

3. Results

Simulations show that the projection error for dual-energy can affect the reconstruction of the iodine mass-thickness in a remarkable way. The error due to the bone layer is varying between 20% (bone thickness 0.5 cm) and 40% (bone thickness 1 cm). Statistical fluctuations are much less important, with respect to the projection error. On the contrary, the triple-energy reconstruction does not show significant projection error. Simulations have shown that triple energy can strongly reduce the projection error, until making it negligible. The comparison of the projection errors of the two methods indicates that triple-energy is able to reduce it from 10 times (background = 100% soft tissue) up to 60 times (background = 100% bone). Figures 2 and 3 show some examples of the simulated images of the phantom at different energies and the reconstructed density maps obtained for the dual and the triple energy methods, respectively.

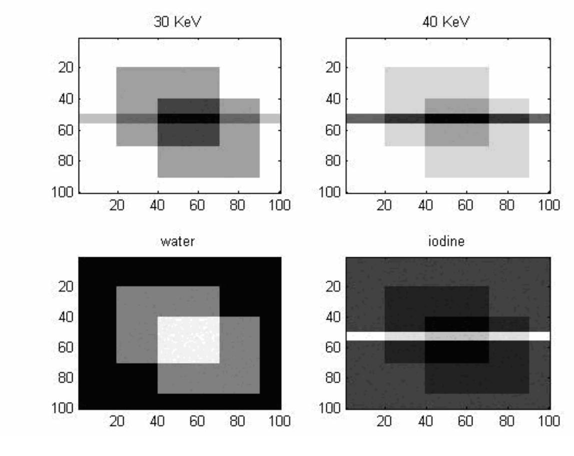

Figure 2: Simulated image of the phantom obtained at 30 keV and 40 keV (top row), and reconstructed density map of basis materials (bottom row) with dual-energy algorithm.

The measurements performed on a reference background (35% Aluminum, 65% water) allowed us to estimated the accuracy and the sensitivity of the two methods. Experimental data demonstrate that, for dual-energy, the projection error is dominant over the statistical noise, also at very high concentration of Iodine (60000 ppm, whereas 1 ppm corresponds to 1 μg of Iodine over a g of background). For triple-energy the projection error is not significant for concentration higher than 900 ppm. The minimal concentration detectable has been estimated in 2400 ppm for dual-energy, whereas for triple-energy it is feasible to detect 660 ppm. In the case of triple energy, that value is close to the statistical error of the measure (± 330 ppm).

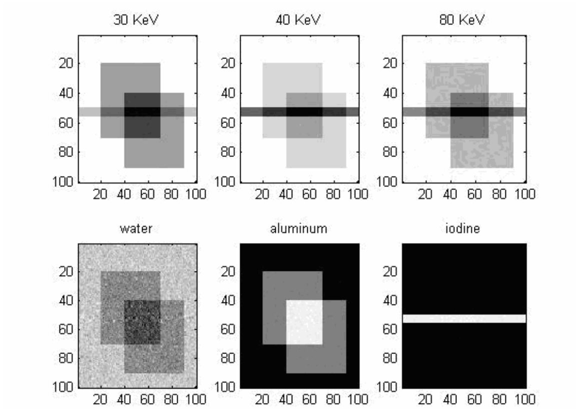

Figure 3 Simulated image of the phantom obtained at 30 keV, 40 keV and 80 keV (top row) and reconstructed density map of basis materials (bottom row) with triple-energy algorithm.

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