Short communication

Single-shot dual-energy subtraction mammography with electronic spectrum splitting: Feasibility

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Abstract

We present a single-shot dual-energy subtraction mammography technique using an energy sensitive photon counting detector. An electronic threshold near the middle of the X-ray spectrum discriminates between high- and low-energy photons, and allows the simultaneous acquisition of high- and low-energy images which can be combined to suppress anatomical clutter. By setting the electronic threshold close to 33.2 keV (the k-edge of iodine) the system is optimized for dual-energy contrast-enhanced imaging of breast tumors. This method eliminates the need for separate exposures which might otherwise lead to motion artifacts. The method is illustrated in phantom images.

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1. Introduction

The neovascularity accompanying tumor growth above a certain size (a few millimeters) often results in increased permeability of the blood vessels [1]. The intravenous administration of iodinated contrast agent will lead to an increased concentration of iodine in and around the tumor. As the attenuation of iodine is much higher than for breast tissue, this will result in more readily visible tumors regardless of any additional techniques applied.

Two additional techniques can be applied to increase detectability and classification of tumors when contrast agents is used in digital mammography: the first is the temporal subtraction method based on first obtaining a pre-contrast image and then one or several post-contrast images. By means of logarithmic subtraction, areas where iodine has accumulated can be identified [2,3] and the difference in kinetics of the enhancement (speed of take-up and wash-out of the contrast agent) can be used to differentiate malignant abnormalities from benign [4].

The second method is contrast-enhanced dual-energy subtraction imaging [5–7]. With this method, two post-contrast images are obtained using different X-ray spectra. By obtaining separate high- and low-energy images of the breast, the difference in attenuation from different glandularity fractions is suppressed by subtracting a fraction of the logarithm of the low-energy image from the logarithm of the high-energy image. The resulting image will preferentially show areas where the uptake of the contrast agent has been intense. Lewin et al. [7] demonstrate that this method, using standard iodinated contrast agent, has the potential to better detect breast cancer tumors in clinical settings. The benefit of the second method over the first is a shorter acquisition time and thus less chance of motion artifacts as the breast does not have to be decompressed (partially released from the compression paddles) while contrast agent is being administered.

Whether temporal subtraction imaging (method one) or dual-energy imaging (method two) is best is an open question, but this technical note focuses on the dual-energy tissue cancellation method applied to contrast-enhanced tumors as described by Lewin et al. [7] However, instead of two exposures using X-ray spectra generated by two different tube voltages, the high-and low-energy images are acquired simultaneously. This can be obtained by having an energy threshold in the photon counting energy sensitive detector around the k-edge of iodine. This eliminates the need for two exposures and reduces the risk of
any motion artifacts stemming from patient movement in the interim period between exposures.

We will briefly present the experimental setup as well as the phantom dual-energy images obtained with a photon counting energy sensitive detector. This illustrates the feasibility of the proposed spectrum-splitting technique.

2. Materials and methods

2.1. Experimental setup

The experimental setup employs a scanned slot geometry with a silicon strip detector. The detector and geometry are described thoroughly elsewhere in literature [8–10]. The detector was fixed and the object was scanned across the X-ray beam using high accuracy step motors. Prior to entering the object, the X-ray beam had been collimated in a 50 μm wide slot which yields the resolution in the scan direction. The resolution in the perpendicular direction is given by the detector pitch which is also 50 μm.

In this experiment, a 128 channel pulse counting application specific integrated circuit (ASIC) was used wire-bonded to the silicon detector. Each channel has a pre-amplifier, a shaper, two discriminators and two counters. The ASIC employs anticoincidence circuits for blocking signals that are shared across the boarder between channels in the silicon detector due to lateral charge diffusion. The ASIC allows 3-bit trimming of each individual channel threshold. The lower threshold (discriminator) level is set to eliminate electronic noise and the second is tuned to eliminate any counts from a 32 kVp spectrum but to count photons of higher energies. This effectively adjusts the electronic threshold to a signal amplitude close to the k-edge energy of iodine.

2.2. Anatomical clutter phantom

The phantom used to simulate anatomical clutter consists of two parts. The first is a 27 mm high PMMA box of 50 mm × 50 mm filled with PMMA cylinders (of height and diameter 6 mm) and olive oil (approximative summation formula C₁₈H₃₆O₂ and specific weight 0.92). Four cylinders are hollow and filled with diluted iodinated contrast agent (iohexol; Omnipaque 300 mg I/ml, Amersham Health, Princeton, NJ). The concentration in the four cylinders range from 0.62 to 1.55 mg I/cm² which are in the lower part the clinically relevant range as used by Jong et al. [4]. The second part of the phantom is a square PMMA box of 15 mm’s height and filled with shattered pieces of PMMA with different and irregular sizes ranging from 1 to 15 mm. The void in this second box was also filled with oil and the boxes were placed firmly together yielding a 42 mm thick phantom. A schematic of the object locations within the phantom and an X-ray image of the phantom is shown in Fig. 1.

The linear attenuation coefficients of the PMMA and olive oil (as calculated from Ref. [11] using the chemical composition above) are presented in Fig. 2 together with linear attenuation coefficients for glandular and adipose tissue (using the compositions in Ref. [12]). Even though the attenuation for PMMA/oil is everywhere lower than for glandular/adipose tissue, the differences $\mu_{\text{PMMA}} - \mu_{\text{oil}}$ and $\mu_{\text{glandular}} - \mu_{\text{adipose}}$ are similar and this difference is what affects the appearance of anatomical clutter the most.
3. Results

The phantom X-ray image was acquired with a 45 kVp tungsten spectrum\(^1\) filtered with 2 mm additional aluminum. The dose was estimated to 1.0 mGy had the phantom been composed of average breast tissue. Two separate images were constructed from the high- and low-energy counters separately. The high- and low-energy images are presented in the top row of Fig. 3 and the lower row presents the summation (total) image and the dual-energy image obtained from logarithmic subtraction:

\[ I = \log I_{\text{high}} - w \log I_{\text{low}}. \]

The weight factor \(w\) is determined to be 0.52 using the method described in chapter 7 of Ref. [13].

Note, that cylinder number 1 is clearly visible in the dual-energy image and that cylinders 2 and 4 are somewhat visible. This is a clear progress over the detectability in the non-processed images.

4. Discussion and conclusion

This technical note is intended to illustrate the feasibility of single-shot dual-energy imaging using electronic spectrum splitting. Several issues remain to be investigated, the most important of which is spectral optimization for maximal dose efficiency. Given the shape of the incident spectra, the dose efficiency depends on the relative high- and low-energy photon fluence. By applying an electronic threshold at the k-edge one is limited to filtering and adjusting kVp to obtain the correct relative high- and low-energy fluence. This limits the possible shapes of the high- and low-energy spectra and the effects of such a constrained optimization warrant further investigation. However, since there is a large discontinuity in the iodine attenuation at the k-edge, having the high- and low-energy spectra concentrated on each side of this energy will be beneficial to the SDNR in the dual-energy image.

Other limitations are of a more technical nature concerning the detector and include capturing the effects of charge-sharing (energy leakage in the detector whereby a high-energy photon interacting close to the boundary of another detector channel is erroneously registered as a low-energy photon), the limited energy resolution and effects of scattered radiation in the detector. When the above mentioned limitations have been modelled, and further experiments carried out with detailed measurements, the effectiveness of this method compared to existing ones can be established.

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References


