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Scanning K-Edge Subtraction Imaging Using Laser-Compton Sources as a Method for High-Contrast and Low-Dose Mammography

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Abstract: A single energy tuning, K-edge subtraction imaging method using laser-Compton sources (LCS) is presented. The narrow spectral bandwidths that LCSs provide can improve clinical dual-energy mammography by 1000x in contrast enhancement or dose reduction.

1. Background and Motivation

There is a critical clinical need to improve mammographic screening. Mammography can miss up to 20% of cancers and reportedly the number of patients who have a positive mammogram that ends up being cancer is as low as 5% [1]. These statistics only get worse the denser tissue a patient has [2], so much that the Food and Drug Administration in the United States now requires disclosure of the unreliability of mammography to patients who have dense breasts. Dual energy methods, like that of contrast enhanced dual energy mammography (CEDEM) have shown to be better than traditional mammograms and as good as magnetic resonance imaging (MRI) in detecting breast cancer [3]. In this paper, we introduce a method that can improve this imaging modality further with a singly energy tuned laser-Compton source (LCS).

2. Laser-Compton Sources and K-Edge Subtraction Imaging

LCSs generate x-rays by taking advantage of Compton scattering. A relativistic electron beam is collided head-on with a laser pulse in a tightly focused spot to increase probability of scattering. The scattered x-rays are highly focused in the direction of electron propagation and the energies are double Doppler upshifted going as the square of the electron energy. These sources have tunable energies, narrow energy bandwidths, and high brilliance [4].

K-edge subtraction (KES) imaging is a contrast-based imaging modality and the one in which dual energy mammography works. By taking two images with energies tuned above and below a contrast element's K-edge energy, subtraction of these two images will result in increased enhancement at the location of the contrast. Since contrast agents tend to pool in tumors, this makes KES a desirable imaging method for cancer screening. The image quality in KES improves the closer the source energies are to the K-edge energy as it: 1) increases the signal due to the discontinuous change in photo-absorption cross section, and 2) decreases noise as the background scattering cross sections become closer in value. Clinical CEDEM sources use the traditional bremsstrahlung x-ray tubes that are broad in bandwidth and have energies tuned far away from the K-edge, making such systems not optimized with much room for improvement. LCS sources, on the other hand, can get very close to the K-edge as they are tunable and have small bandwidths, as small as 0.1% ΔE/E full width at half maximum (FWHM) for properly designed systems. Figure 1 compares a 1.5% bandwidth (BW) LCS source with a clinically used CEDEM source for iodine-based KES mammography.

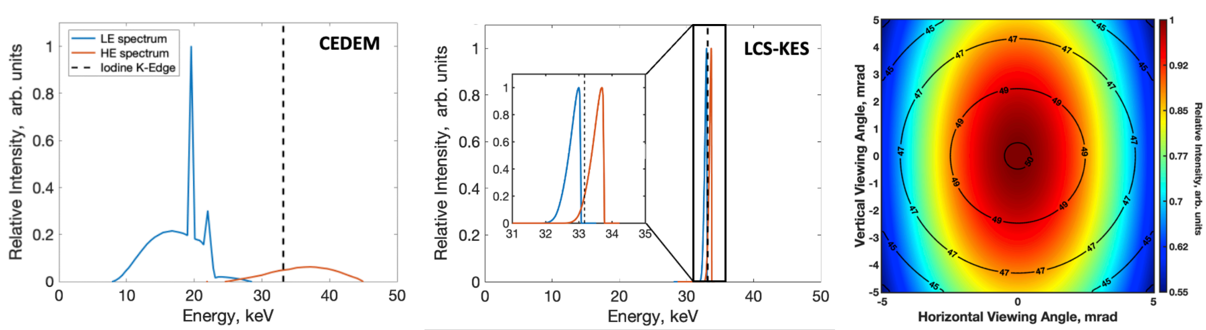

Fig. 1. (left) GE Senobright clinical CEDEM source spectra. (middle) 1.5% BW dual-energy LCS spectra. (right) LCS energy-intensity-angle correlation spectrum.

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Source switching, like in CEDEM, energy switching of a dual-energy LCS, for KES imaging takes time. However, due to the Klein-Nishina scattering probabilities, LCSs have an angle correlated energy-angle spectrum. This enables tuning of a LCS to a single electron beam energy that produces an x-ray spectrum that contains photon energies that are both above and below a contrast element's K-edge at the same time separated in space. By scanning an object through the beam, one can effectively reconstruct a KES image without changing any beam parameters, making a faster time image acquisition scenario [5,6].

3. Computational Experiments and Results

Simulated imaging experiments were performed to verify the feasibility of scanning LCS KES mammography and were compared to General Electric's Senobright system, a commercially available and clinically used CEDEM source, as well as a dual energy LCS KES mode that we have investigated previously [7]. The VICTRE computation breast phantom dataset was used to randomly generate phantoms of various morphologies that are anatomically and density accurate. Four different breast densities were used to represent the 4 BI-RADS density classifications. Tumors were randomly placed within the phantom at the terminal duct lobular units, which is the most common sites for cancer to form. The concentration of iodine was varied between 0.1% (weakly enhancing) to 0.5% (strongly enhancing) tumors. The signal-to-noise as well as relative contrast were compared between all the imaging modalities.

Figure 2 shows image comparisons between the three imaging modalities studied for a dense phantom using strongly enhancing tumors. The relative contrast for Senobright, dual energy LCS, and scanning LCS were 0.031, 38.87, and 39.91 respectively. Between the Senobright and the LCS images, there is an over 1000x increase in the contrast indicating a remarkable improvement in the signal. Since the Senobright already improves upon traditional mammography and is as sensitive as MRI, one could potentially obtain the same contrast image as CEDEM in an LCS using only 0.1% of the dose. The scanning LCS image produces the same image quality as a direct energy tuning image due to the similar bandwidth spectrum applied indicating the same quality image as direct energy tuning can be obtained without time consuming energy switching. There is, however, increased noise around the edge of the scanning LCS image compared to the dual energy LCS image due to lower flux of photons on the edge of the spectrum.

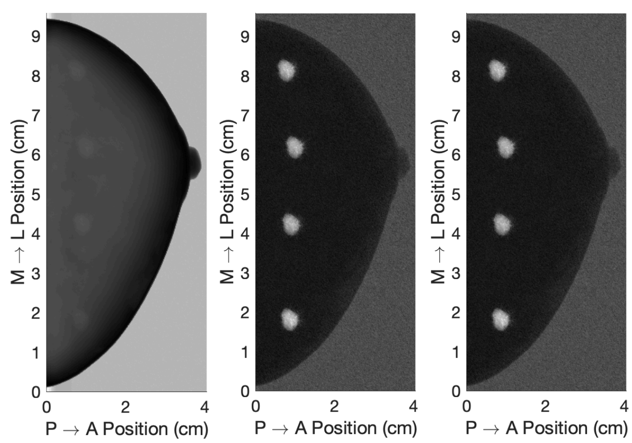

Fig. 2. Preliminary simulation KES images of a dense breast computational phantom using Senobright (left), dual energy LCS (middle) and single energy scanning (right) sources.

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