Dose in Dual Energy CT: Does It Go Up or Down?

MAYO CLINIC

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INTRODUCTION

Dual energy CT was first proposed by Godfrey Hounsfield in 1973 [1]. Subsequent investigations led to the development of a commercial CT system with dual energy capabilities [2-4]. Due to limitations in CT technology at that time, dual energy CT was abandoned, returning in 2006 with the introduction of a scanner geometry known as dual-source CT. Designed for cardiac applications, the system (SOMATOM Definition, Siemens Healthcare) used two tube/detector array pairs to improve temporal resolution by a factor of two relative to a single source scanner [5]. The dual source design also enabled the simultaneous acquisition of dual energy CT data [6]. Alternate strategies for dual-energy data acquisition include a rapid kV switching approach similar to that used in the 1980's, and a multi-layer ("sandwich") detector design that gathers the low and high energy data from two different detector layers that are separated by an additional filter layer.

This educational exhibit focuses on the dose requirements for dual-source, dual-energy CT (Figure 1) relative to current single energy clinical CT protocols to answer the question:

Dose in Dual Energy CT (DECT): Does It Go Up or Down?

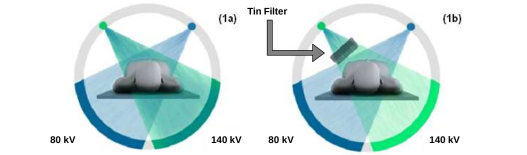

Figure 1: Schematic of the second generation dual source scanner without (1a) and with (1b) an additional flat tin filter in the 140 kV x-ray path

BACKGROUND

Dose reduction methods for both Single Energy and Dual Energy CT

Modern CT scanners utilize a variety of technologies to reduce patient dose levels. These include both hardware and software approaches. While there are many measures that can be taken to reduce radiation dose, the following are particularly relevant for the dual-source CT systems discussed here.

Automatic Exposure Control (AEC)

AEC systems use patient-specific information to adjust the tube current according to the amount of patient attenuation in the scan plane. The evaluated systems estimate patient attenuation using the CT radiograph (e.g. topogram) taken to plan the scan range, as well as the real-time attenuation measurements taken during the scan. The tube current is adjusted to deliver the lowest dose necessary to achieve the level of image quality prescribed by the operator. The evaluated systems automatically adjust the tube current in both the angular- and z-axis directions. AEC is applied to each tube individually and simultaneously [7].

Adaptive Dose Shield (Flash Scanner only)

Conventional spiral CT techniques require the irradiation of anatomy outside of the desired scan volume in order to reconstruct images at the start and end of the scan range, an effect referred to as over-scanning. New generation dual source scanners utilize a hardware-based solution in which the x-ray beam collimation is increased from zero, at the edge of the scan range, to the total nominal beam width, once fully inside the scan range. When the scan position reaches one beam width from the end location, the collimator opening is decreased, reaching zero at the end of the scan range. The dose reduction is greatest for short scan lengths or high pitch values [8].

Beam Filtration

All CT scanners use filters in the x-ray beam to selectively decrease unwanted photon energies. These filters can be varied for specific scan regions and/or patient sizes, for example head or cardiac regions, or pediatric patients.

Dose reduction methods specific to Dual Energy CT

Tin Filter (Flash Scanner only)

Dual source CT allows independent control of both the tube current and tube potential for each x-ray tube. Thus, at low kV settings, an increased mA can be used to maintain acceptable noise levels in the low energy image. Figure 2 provides an illustration of the x-ray beam spectra at 80 and 140 kV, as well as the level of overlap between the two spectra. Kelcz demonstrated that the ability of dual-energy CT to differentiate between materials (e.g. iodine and bone) depends on the separation between the high and low energy spectra [3]. To increase the separation between the 80 and 140 kV beams shown in Figure 2b, a tin filter was introduced into the high-energy beam (Figure 1b) [9,10]. Figure 2c illustrates the increased separation achieved with use of the tin filter.

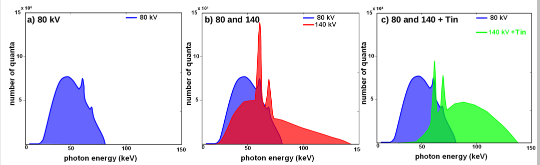

Figure 2: Illustration of the x-ray tube spectra for (a) 80 kV, (b) 80 and 140 kV, and (c) 80 and 140 kV + Tin. The spectra in 2c have a significantly reduced overlap compared to 2b.

DOSE vs. NOISE

We evaluated the relationships between dose and noise for three phantom sizes (20, 30, and 40 cm) and compared this with single-energy 120 kV scans of the same phantoms [11]. The 20 and 30 cm phantoms were scanned using 80 and 140 kV (80/140 kV), with and without the tin filter, and the 30 and 40 cm phantoms were scanned with 100 and 140 kV (100/140 kV), with and without the tin filter.

Measurements were made - prior to the availability of the Definition Flash, which has inherent tin filtration - on a modified Definition Dual Source scanner with an additional 0.4 mm flat tin filter for the high-kV tube [9,10]. The composition ratios (C_{ratio}) used to produce the linearly-mixed images were set at 0.3 for 80kV/140kV, 0.4 for 100kV/140kV, and 0.6 for 100kV/140kV +Tin, where the mixed image is a linear combination of the low and high kV images:

Mixed image=Low-kV image × C_{ratio} + High-kV image × (1- C_{ratio}).

These data (Figure 3) were used to find the dose (in terms of Volume CT Dose Index, CTDIvol) where image noise in the DECT images matched that of our clinical single-energy abdomen protocol (120 kV).

An American College of Radiology (ACR) image quality phantom was scanned in the dual energy and single energy modes (Figure 4) at the dose levels predicted to yield comparable image noise. The composition ratio used to produce the linearly-mixed images for this portion of the study were set to 0.5 for all acquisition modes. This 0.5 composition ratio setting reflects the default values on the newer generation dual-source scanner (Definition Flash)

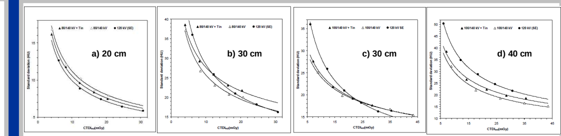

Figure 3. Noise vs. scanner output ($CTDI_{vol}$) for 20, 30 and 40 cm diameter water phantoms:
(a) 20 cm phantom using 120 kV (●), 80/140 kV with tin (▲) and 80/140 kV without tin (△);
(b) 30 cm phantom using 120 kV (●), 80/140 kV with tin (▲) and 80/140 kV without tin (△);
(c) 30 cm phantom using 120 kV (●), 100/140 kV with tin (▲) and 100/140 kV without tin (△); and
(d) 40 cm phantom using 120 kV (●), 100/140 kV with tin (▲) and 100/140 kV without tin (△).
The solid lines represent a fit of the data to a power-law curve.

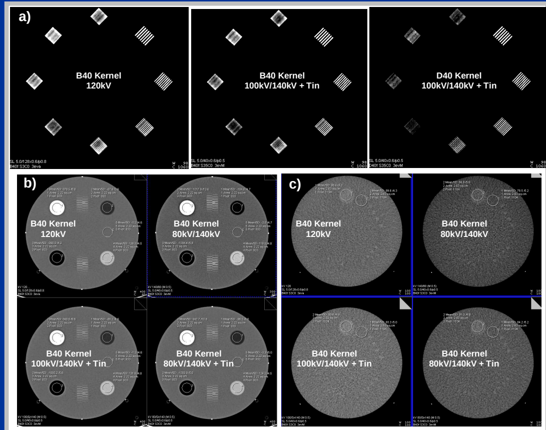

Figure 4: Images of the (a) high spatial resolution module, (b) CT number module and slice width module and (c) the low contrast module of the ACR CT Accreditation image quality phantom

RESULTS

The data in Figure 3 demonstrates that the single energy scan mode provides the lowest dose for the same image noise level only for the 20 cm phantom. For the 30 and 40 cm phantoms, the dual energy modes can achieve the same level of image noise in the mixed image at a lower dose level (CTDIvol). Comparing the dual energy modes with and without the tin filter, use of the tin filter reduced the required dose for the 20 cm phantom, had minimal effect on dose for the 30 cm phantom, but increased the required dose for the 40 cm phantom.

RESULTS

Figure 4 compares images from scans performed with equal doses using the following modes: 120 kV, 80/140 kV, and 80/140 kV + Tin, and 100/140 kV + Tin. Comparing the single energy 120 kV scan with the 100/140 kV + Tin (using medium smooth body kernels), no evidence of image degradation is seen in the dual energy images. The CT numbers for the 120 kV and 100/140 kV + Tin images are very similar. For materials with high atomic number (e.g. bone), the CT numbers with 80/140 kV are higher than at 120 kV. This can actually be of benefit in visualizing small hyper-attenuating lesions. No difference in low contrast resolution was observed. Measurements taken in the uniformity/noise/distance accuracy module (not shown) also demonstrated no differences between the acquisition modes.

DISCUSSION

When data are acquired and reconstructed with factory (default) dose partitioning and blending ratios (C_{ratio}), the noise levels for DECT mixed images differed from that of single energy images when the same CTDIvol was used. Noise in the DECT mixed images was higher than in the single energy images for the small (20 cm) phantom, which represents a very tiny person. For the more realistic attenuation levels of an adult, i.e. the medium (30 cm) and large (40 cm) phantoms, noise was lower using DECT mixed images relative to 120 kV single energy, which would allow a potential dose decrease in DECT without impacting image quality.

It is important to note that DECT provides the flexibility to alter the noise level by adjusting the linear blending ratio (C_{ratio}) used to combine the low and high energy images. Non-linear sigmoidal blending techniques could also be used to improve iodine conspicuity.

We further demonstrated that image quality in single energy (120 kV) and DECT mixed images is comparable on the systems evaluated. Based on these results, we conclude that:

At equal dose, the mixed images from 100/140 kV + Tin dual energy scans have very similar noise, CT numbers and overall image quality to images from a 120 kV single energy scan. The 80/140kV and the 80/140kV + Tin scan modes created images with CT numbers differing from the single energy images for high Z material (e.g. bone). However at the same dose, noise and other important image quality characteristics were very similar. Thus, DECT scans can be used, for example, to automatically segment bone from iodinated vessels and to identify kidney stone composition in vivo, at doses similar to a single energy scan, without compromising overall image quality. Rather, in some cases, iodine enhancement was improved relative to 120 kV.

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