ELSEVIER

Seminars in
ROENTGENOLOGY

Advances in Computed Tomography in Thoracic Imaging

**Asha Kandathil, MD, Fernando Kay, MD, Kiran Batra, MD,
Sachin S. Saboo, MD, and Prabhakar Rajiah, MD**

Introduction

Computed tomography (CT) has become an integral imaging modality in the evaluation of thoracic disorders. Currently available multi-detector CTs with volumetric scanning permits 3-dimensional imaging of the chest with multiplanar reconstruction and volumetric display. Rapid scanning with wider coverage is now possible with the use of wide-array multi-detector scanners as well as high pitch helical mode of dual-source CT scanners, which enables motion-free images. Advances such as iterative reconstruction algorithm enable acquisition of high-quality images at low radiation doses. Dual-energy techniques enable material and tissue characterization beyond what is possible with a conventional CT scan. Quantification techniques have become sophisticated, including quantification of chronic obstructive pulmonary disease (COPD) and interstitial lung disease. Dynamic imaging and virtual bronchoscopy is available in the evaluation of airways. Advances in intervention include the use of CT in electromagnetic navigational bronchoscopy and image guided video assisted thoracoscopic surgery (VATS). Computer aided detection (CAD) systems have been applied to thoracic CT to improve the accuracy and efficiency of lesion detection and characterization. Textural analysis provides quantitative information of gray-level patterns, pixel interrelationships, and spectral properties of an image. Quantitative image features can be correlated to tumor phenotypes by radiomics and with molecular genotypes in radiogenomics. In this article, we review these advances in CT in thoracic imaging.

Dual-Energy CT

Dual energy CT (multi-energetic CT or spectral CT) has the ability to characterize materials or tissues beyond what is possible with a conventional CT. Attenuation coefficients of an object obtained at 2 distinct polyenergetic X-ray beams with different energy levels provide sufficient information to describe the image pixels as combinations of two materials, with known linear attenuation coefficients, which in turn provides estimates of material concentration per pixel.[1,2] Advanced mathematical algorithms may be used to decompose objects into 2 or even more materials.[3] There are several dual-energy CT (DECT) technologies for acquiring 2 sets of X-ray attenuation profile at high and low energy levels (Figure 1). The simplest strategy is the *rotation-rotation* technique, in which 2 acquisitions, one low and the other high energy are consecutively obtained from a single source CT scanner at the same table position.[4,5] This is not commonly performed due to patient motion and differences in contrast phase between two subsequent acquisitions. *Dual spin technology* is a variation of this method, performed in a volumetric scanner, where the entire area can be scanned in 1 gantry rotation, with 2 different energies at 2 consecutive rotations. *Dual-source CT* scanners rely on a set of two x-rays tubes built at a 90° offset angle in a single gantry, allowing image acquisition at 2 different energy levels (Figure 1A).[6] In *rapid kVp switching* technology, the kVp of the x-ray projection is rapidly switched from a low to high energy (Figure 1B).[7-9] *Split beam* technology uses a filter to divide the x-ray beam into high and lower energies. *Dual-layer technology* is a detector-based technology, with the top layer of detector absorbing low-energy photons and the bottom layer absorbing high-energy photons (Figure 1C).[10] *Photon counting* technology is able to discriminate multiple energy levels, but is not clinically available.

In addition to conventional images (combined or blended or mixed or routine diagnostic images), multiple spectral images are generated in a DECT scanner. This includes virtual noncontrast, iodine map, virtual monoenergetic image (VMI)

Cardiothoracic Imaging, Radiology Department, UT Southwestern Medical Center, Dallas, TX.

Address reprint requests to Prabhakar Rajiah, MD, Department of Radiology, Cardiothoracic Imaging, UT Southwestern Medical Center, E6.120 B, Mail Code 9316, 5323 Harry Hines Boulevard, Dallas, TX 75390-8896. E-mail: Prabhakar.Rajiah@utsouthwestern.edu

https://doi.org/10.1053/j.ro.2018.02.006

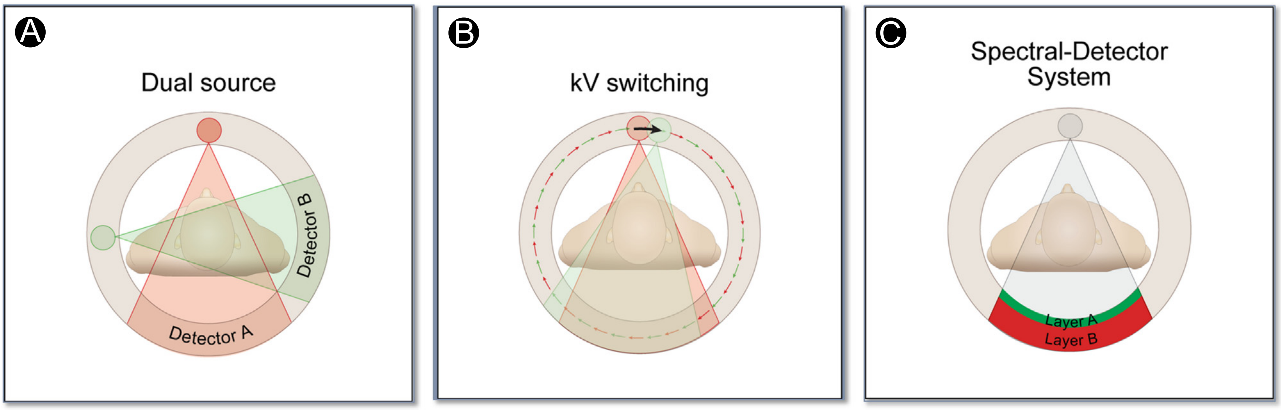

Figure 1 Dual energy CT technologies. (A) Dual source CT with 2 x-ray tubes operated at different tube potentials. (B) Rapid kVp switching technology with the kVp switched between high and low tube potentials for each x-ray projection. (C) Dual-layer spectral detector CT system with the top layer absorbing low energy photons and the bottom layer absorbing high energy photons. (Color version of the figure available online.)

and effective atomic number-based image. VNC images are derived by subtraction of iodine from contrast-enhanced images.[11-14] CT numbers measured on VNC have shown good agreement with those on true noncontrast (TNC) images with exception of values in fat tissues, which tended to be overestimated by VNC,[15,16] with best performances of dual-source CT, followed by rapid-kVp-switching, and dual-layer detector approaches.[17] Iodine maps demonstrate pixels containing iodine, with quantification of iodine in mg/mL (Figure 2). VMI are images that mimic a monoenergetic image obtained at a particular energy level (usually from 40-200 keV), obtained by mathematical processing and linear combination of basis images.[18] VMI at high energy levels are useful in decreasing several image artifacts, while VMI at low energy levels are useful in enhancing vascular contrast and improving lesion conspicuity[19] (Figure 3). Effective atomic number-based images are color coded based on the atomic number of tissues or materials.

Iodine imaging by DECT has been used as a surrogate method to assess perfusion.[20,21] However, it estimates perfusion at a single-time point unlike dynamic CT or MR perfusion, which use a time-intensity curve of contrast medium bolus first-pass to derive perfusion parameters.[22] Consequently, DECT perfusion is dependent on acquisition timing and hemodynamics. Although iodine maps (Iodine$_{[\text{no water}]}$ images) are generated by DECT material decomposition algorithms, and provide estimates of iodine concentration per pixel,[23] pulmonary blood volume (PBV) images derives from a calculated ratio of the mean enhancement of the pulmonary parenchyma with a reference vessel (usually the pulmonary trunk), further corrected by a calibration factor (Figure 4).[22] PBV image has been used to detect perfusion defects associated with pulmonary emboli, with sensitivities and specificities of 75% and 80% per patient and 83% and 99% per segment, respectively compared to perfusion scintigraphy[24] (Figure 5). Another study showed sensitivity and specificity of 77% and 98%, respectively of iodine imaging compared to SPECT or CT.[25] Perfusion defects in iodine maps correlated with obstructive emboli in 82% of positive cases in a larger study.[26] Combination of DECT iodine imaging with CT angiography images have shown the highest performance for the diagnosis of pulmonary embolism in an animal model.[27] DECT also

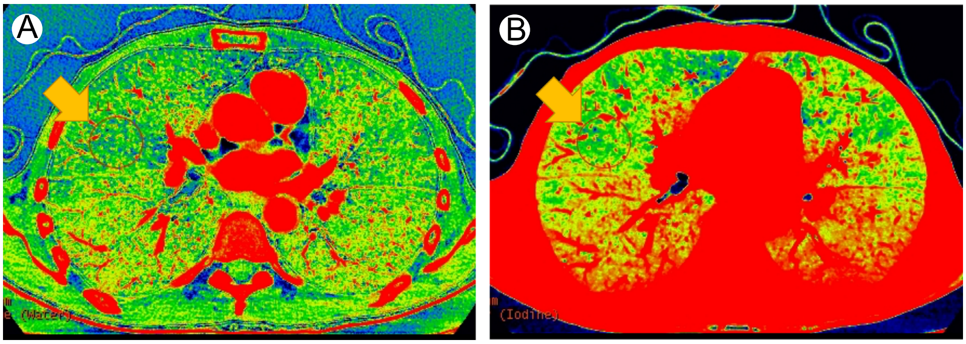

Figure 2 Dual-energy CT material decomposition. CT angiography acquired for the assessment of pulmonary embolism on a rapid kvp switching dual-energy CT scanner. By using a material decomposition algorithm, it is possible to estimate the concentration of at least 2 materials with known linear attenuation coefficients. In this example, iodine and water were selected as the material pair. (A) Iodine-based and (B) water-based images. By drawing a region of interest (arrows), it is possible to estimate the concentration of iodine (A) and water (B) (for example, 5.5 and 187 mg/cm^3, respectively), assuming that pixels are only composed by either one of the materials. (Color version of the figure available online.)