Dual-Energy CT: General Principles

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OBJECTIVE. In dual-energy CT (DECT), two CT datasets are acquired with different x-ray spectra. These spectra are generated using different tube potentials, partially also with additional filtration at 140 kVp. Spectral information can also be resolved by layer detectors or quantum-counting detectors. Several technical approaches—that is, sequential acquisition, rapid voltage switching, dual-source CT (DSCT), layer detector, quantum-counting detector—offer different spectral contrast and dose efficiency. Various postprocessing algorithms readily provide clinically relevant spectral information.

CONCLUSION. DECT offers the possibility to exploit spectral information for diagnostic purposes. There are different technical approaches, all of which have inherent advantages and disadvantages, especially regarding spectral contrast and dose efficiency. There are numerous clinical applications of DECT that are easily accessible with specific postprocessing algorithms.

Keywords: CT detector technology, dual-energy CT, dual-source CT, rapid voltage switching, spectral CT

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The term "dual-energy CT," or "DECT," refers to CT that uses two photon spectra; therefore, DECT is sometimes also referred to as "spectral CT." In clinical practice today, two different spectra are generated either by switching the voltage of one x-ray tube or by running two tubes at different voltages, and spectral information is gained from two absorption measurements with normal CT detectors. On the other hand, in the near future, energy-resolving detectors may be capable of resolving spectral information at two or more energy levels using the polychromatic spectrum of one x-ray tube.

X-Ray Spectra

The sources that can provide a sufficient output of quanta for diagnostic imaging are tubes with rotating anodes that have polychromatic spectra consisting of a continuous spectrum of bremsstrahlung superimposed with characteristic lines of the tungsten material of the anode (Fig. 1A). Thus, there are not two distinct photon energies as the term "dual-energy CT" suggests but, rather, two x-ray spectra. The maximum energy of the photons is defined by the voltage, whereas the mean energies are significantly lower and their differences smaller than one may expect—for 76- and 56-keV mean energy, respectively, for 140 and 80 kV with typical filters. If two separate tubes are used, an adaptation of tube currents is recommended to obtain a similar total photon output from both sources (Fig. 1B). The settings of 80 and 140 kV are commonly used because they provide the maximum difference and least overlap between the spectra with standard tubes (Fig. 1C). A tube voltage lower than 80 kV is not generally available and is not useful because too many of the quanta would be absorbed by the human body, at least in adults, and 100 kV is preferable in clinical practice for the trunk of the body in patients weighing more than 80 kg. New tube technology offers 70 kV at high currents, which may be of interest for DECT of the extremities or of children (Fig. 1E). Values higher than 140 kV are generally not available with today's tubes but could offer greater spectral contrast. Additionally, filters can be used to rid the high-energy spectrum of low-energy quanta and decrease the overlap with the low-energy spectrum.

In second-generation dual-source (DSCT), a tin filter of 0.4-mm tin improves spectral separation (Fig. 1D). A thicker filter further hardens the spectrum but absorbs so much of the output of quanta that the remainder gets too small even with a maximum tube current; an example is shown for a 1-mm-thick stan-

num filter at 140 kV in Figures 1A and 1B. Still, in the near future, higher tube voltages in combination with thicker filters may offer further improvements in spectral contrast.

Detector Technology

Today's CT detectors integrate all the fluorescent light intensities induced by the impact of photons in the scintillator during a readout interval but do not give account of their energy. Therefore, current DECT approaches either rely on entirely separate x-ray sources and corresponding detectors or rely on reading out the projection data at different time points. Two-layer or "sandwich" detectors with different spectral sensitivities could provide spectral information in single-source systems but to date have not been implemented in clinical scanners. In the future, cadmium-based semiconductors, such as CdZnTe, may serve as semiconductors for photon-counting detectors, which resolve the energy of each individual photon, a method already used in nondestructive material testing and luggage scanners at airports. However, this detector technology cannot yet cope with the high photon flux and cannot provide the high image quality required for clinical CT.

Tissue Properties and Contrast Material

To allow a differentiation on DECT, the tissue or contrast material in the examined area must have spectral properties—that is, differences in x-ray attenuation at different photon energies. The x-ray attenuation is caused by the Compton effect, coherent scatter, and the photoelectric effect. Of these properties, the photoelectric effect strongly depends on the atomic number of the material. Because the photoelectric effect strongly increases with atomic number, most of the atoms in the human body (i.e., hydrogen, carbon, nitrogen, and oxygen) have a rather weak photoelectric effect. Some ions in the body such as calcium or magnesium have a somewhat stronger effect, whereas the photoelectric effect of contrast material (i.e., iodine, barium, or xenon) is very strong. These differences are caused by the interaction of photons with electrons in the shell of the atom. As a result of the photoelectric interaction with the inner K shell, the photoelectric effect peaks at element number 55 (cesium). The element numbers of typical contrast media like iodine (53), xenon (54), and barium (56) lie in this area, so that these elements provide a strong photoelectric effect. In heavier atoms, the second L shell prevails, resulting in a lower photoelectric effect. Therefore, a strong spectral contrast can be achieved on DECT between the light atoms of the body tissues and the heavy atoms of contrast materials. Thus, iodine as a well-established standard contrast material for CT, offers optimal properties for DECT [1–3]. This effect is well known to radiologists because iodine enhancement is obviously much stronger with low tube voltages (e.g., in CT angiography) [4]. With DECT, this difference in spectral behavior can be used to detect and measure iodine on CT images.

Technical Approaches

At present, there are five approaches to DECT: sequential acquisition, rapid voltage switching, DSCT, layer detectors, and energy-resolving or quantum-counting detectors; currently, only the former three are commercially available.

Sequential Acquisition

The approach that requires the least hardware effort is the sequential acquisition of two datasets at different tube voltages. Sequential acquisition can be achieved either as two subsequent helical scans or as a sequence with subsequent rotations at alternating tube voltages and stepwise table feed. This approach may make sense in systems with broad detectors, but a disadvantage is the rather long delay between both acquisitions. The delay is too long to preclude artifacts from cardiac or respiratory motion or changes in contrast material opacification. However, sequential acquisition should be a viable option for clinical DECT applications without contrast material, such as metal artifact removal or kidney stone differentiation.

Rapid Voltage Switching

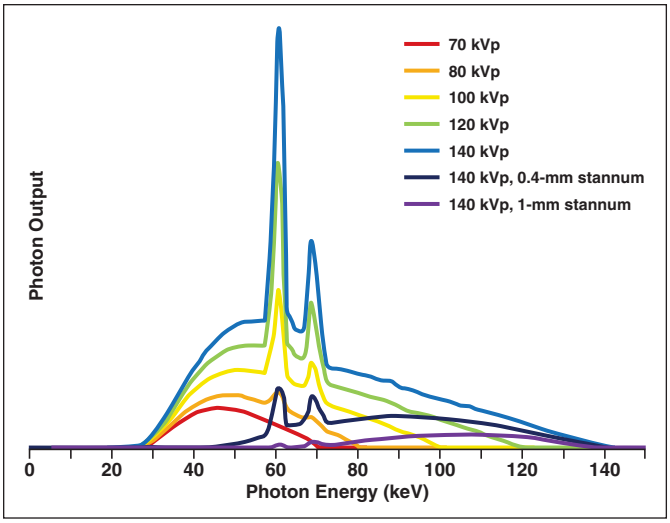
Another approach with very little technical effort is rapid voltage switching (Fig. 2A). With this method, the tube voltage alternates between a high value and a low value, and transmission data are collected twice for every projection or, in practice, for immediately adjacent projections. The rotation speed of the system must be reduced to account for the acquisition of these additional projections and the rise and fall times of the voltage modulation. Therefore, the gantry rotation time usually must be 0.5 second or longer, which prolongs the acquisition time. Another important disadvantage of this setup is the limited photon output at low voltages, which results in high noise and the necessity to choose a relatively high current and correspondingly high dose. Optimally, the tube current should be adapted to the tube voltage to achieve a similar output of photons at both voltages [5]. However, this adaptation of tube current is technically challenging or may even be impossible because the free electrons of the cathode do not become available rapidly enough to accommodate such fast changes in current. Therefore, another approach is to acquire two low-voltage projections for each single high-voltage projection to balance the number of photons available at the two energy levels to some degree [6]. Still, the spectral resolution of rapid voltage-switching systems remains limited and usually requires additional dose because other dose reduction features, such as tube current modulation or an optimized filtration of each spectrum, are not possible.

Dual-Source CT

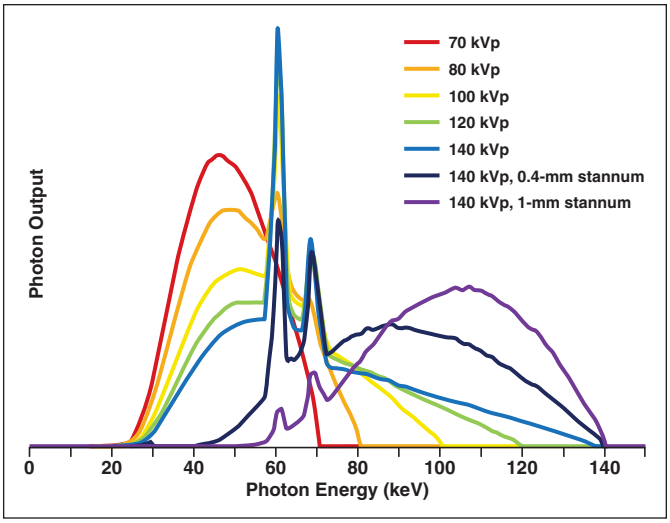
A straightforward approach is DSCT with two tubes running at different voltages and corresponding detectors mounted orthogonally in one gantry (Fig. 2B). This setup requires nearly twofold investments in hardware but offers important advantages for DECT: The voltage, current, and filter can be chosen independently for both tubes to achieve an optimal spectral contrast with sufficient transmission and the least overlap. The data are acquired simultaneously by both orthogonal systems. Although there is an angular offset between both spiral paths, there is no temporal offset in data acquisition because equivalent *z*-axis positions are scanned at the same time. For two detectors to be integrated in one gantry, one of the detectors must be somewhat smaller than the other, technically resulting in an FOV of 33 cm in diameter. In clinical practice, this FOV is sufficient to cover all the vessels and organs in most patients. If some adipose tissue of obese patients is outside the FOV of the dual-energy acquisition, this is generally not of clinical relevance, and the anatomy is covered by the wide FOV of the larger detector. A disadvantage of the orthogonal setup is cross-scatter radiation, which partially hits the noncorresponding orthogonal detector and requires correction. However, in the latest DSCT systems, there are specific detector elements to measure and correct cross-scatter radiation.

Layer Detector

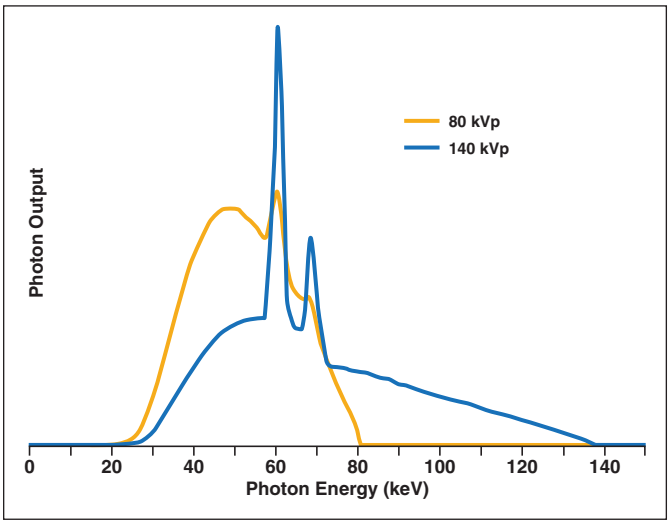
Another approach that is currently not commercially available uses an energy-resolving



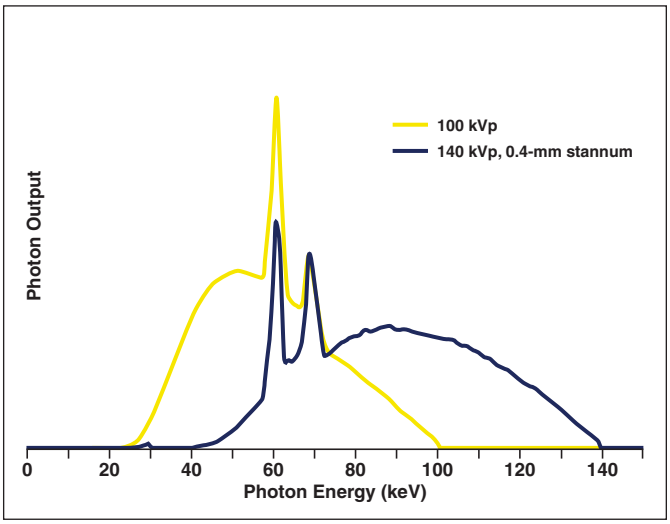
A



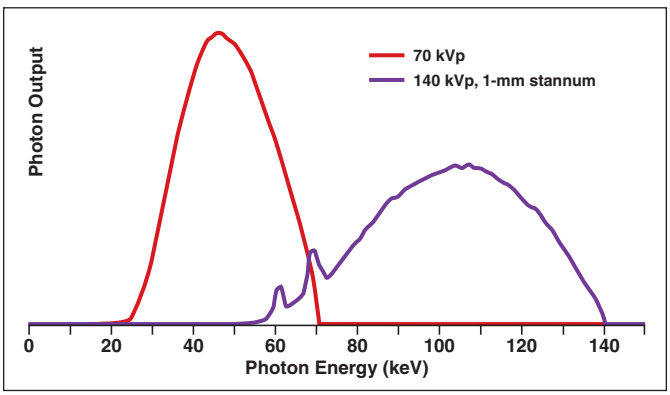
B



C



D



E

Fig. 1—X-ray spectra–simulated spectra based on Monte Carlo techniques.
A, Spectra of x-ray tube at 70, 80, 100, 120, and 140 kV with 0.9-mm titanium and 3.5-mm aluminum filter and at 140 kV with additional 0.4- or 1-mm tin filter. Peaks represent characteristic lines of tungsten anode, and continuous spectrum is result of bremsstrahlung. At equal currents, tube efficiency is optimal at 140 kVp, whereas photon output is significantly reduced in filtered spectra and at 70 kVp.
B, Tube currents can be adapted to achieve equal total photon output. However, this adaptation can be limited by maximum tube current, especially regarding 70- and 140-kVp spectra with 1-mm stannum filter.
C, Combination of 140 and 80 kVp is used in rapid kilovoltage-switching systems for sequential acquisition and in dual-source CT systems. Overlap between both spectra is quite broad, limiting spectral contrast.
D, Combination of 100-kVp and filtered 140-kVp spectra provides improved spectral contrast. Advantage of this combination is that average of both matches 120-kVp spectrum exactly, so that resulting density (in Hounsfield units) can be interpreted like it would be in routine examinations. For photon output to be sufficient, two separate rotating envelope tubes are required, both running at rather high currents.
E, In theory, 70-kVp spectrum and 140-kVp spectrum with 1-mm-thick tin filter could provide excellent spectral contrast. However, in practice, this combination would have too little transmission in human body and would wear out tubes with very high currents.

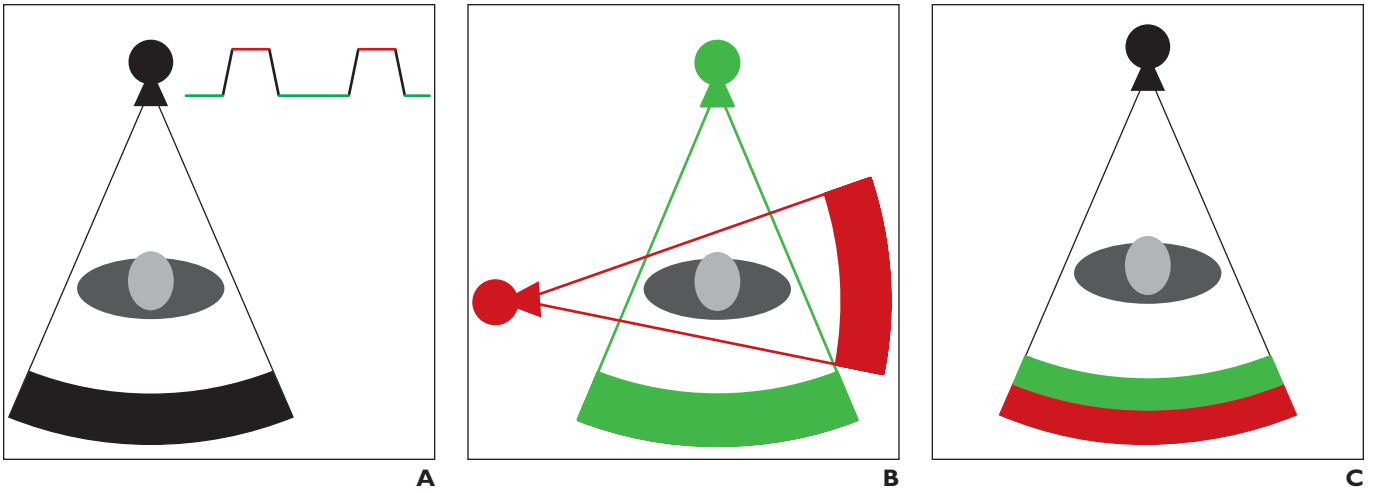


Fig. 2—Technical approaches.
A, Sketch of rapid kilovoltage-switching system containing only one tube and one detector. Voltage is switched rapidly between two levels.
B, Sketch of dual-source CT system with two tubes and detectors mounted orthogonally in one gantry. Tubes are operated at different tube voltages (e.g., 80 and 140 kV). Additionally, filter can be applied to rid high-energy spectrum of low-energy quanta.
C, Sketch of layer detector system with one x-ray tube running at constant voltage. Dual-energy information is derived from two layers of detector with different sensitivity profiles.

detector with the polychromatic spectrum of one tube (Fig. 2C). In a layer detector, the sensitivity of two layers is determined by the scintillator material—for example, consisting of ZnSe or CsI in the top layer and Gd_2O_2S in the bottom layer. With this setup, the scintillator materials determine the spectral resolution, and the sensitivity profiles of the available materials have a rather broad overlap. Therefore, the contrast of the spectral information is limited or requires a relatively high additional dose.

Quantum-Counting Detector

Quantum-counting detectors using, for example, CdZnTe can in principle resolve the energy of each individual impacting photon. This technology can be used to differentiate more than two photon energies and is very quantum-efficient [7]. However, these detector materials get saturated rather quickly, resulting in a rapid drift of the measured signal. Therefore, to date, these detectors work in scanners that are used to scan small animals [8] but cannot handle a photon flux required for clinical CT [9].

Radiation Exposure

The radiation exposure required for DECT depends on the technology used. Generally, the aim is to use the same dose as would be used for a single-energy examination. Only then, it is easily possible to replace standard protocols with dual-energy examinations because an additional diagnostic value is offered without additional dose. Investigations by Ho et al. [10] showed 2–3 times higher doses for DECT based on a single-source system using rapid voltage switching. However, their setup contained a normalization of neither image noise nor dose, so the lower energy spectrum was obtained with the same tube current–time product as the single-energy scan. In DSCT, the tube currents can be tailored so that the dose from both tubes matches that of a routine single-source CT protocol [11]. Comparative studies with external validation using thermoluminescent detectors in Alderson phantoms on a clinical system showed similar or improved contrast-to-noise ratios (CNRs) for DECT at equivalent dose [12].

A direct comparison of the different DECT approaches should include both spectral contrast and dose optimally quantified as CNR per dose. Detailed studies based on Monte Carlo simulations compare different scanner setups. Currently, a maximum spectral contrast is achieved with a DSCT system with optimized voltage, current, and filtration. Regarding the spectral contrast of this DSCT system for iodine-calcium separation in a 30-cm-diameter phantom as 100%, relative CNR per dose ranges between 22% and 45% for layer detector systems [13], around 35% for realistic

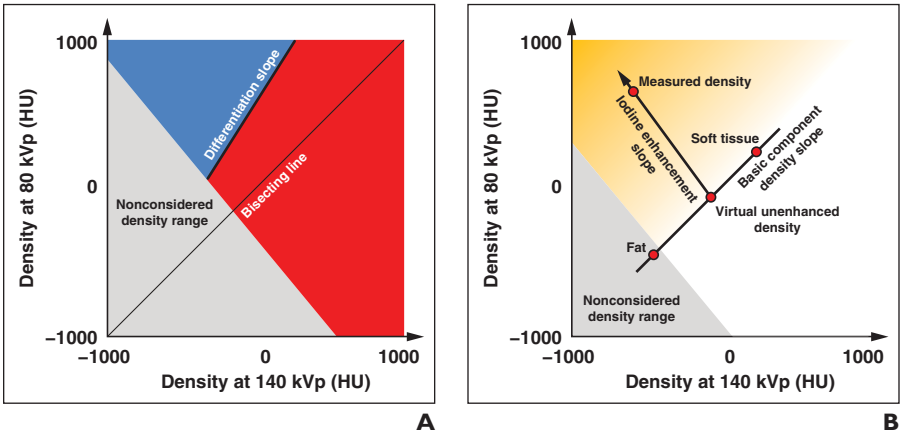


Fig. 3—Diagrams of postprocessing algorithms.
A, Algorithm is used to differentiate two materials from one another. Slope defines separation based on difference in densities of two materials in Hounsfield units. Two materials are color-coded in red or blue (Fig. 4E).
B, Algorithm is used for decomposition of three materials. Two basic components define one slope, and inclination of second slope is defined by photoelectric effect of iodine. Iodine content can be quantified and color-coded by displacement of measured pair of density values along second slope (Figs. 4G and 4I). Virtual unenhanced density is represented by intersection of both slopes (Fig. 4H).

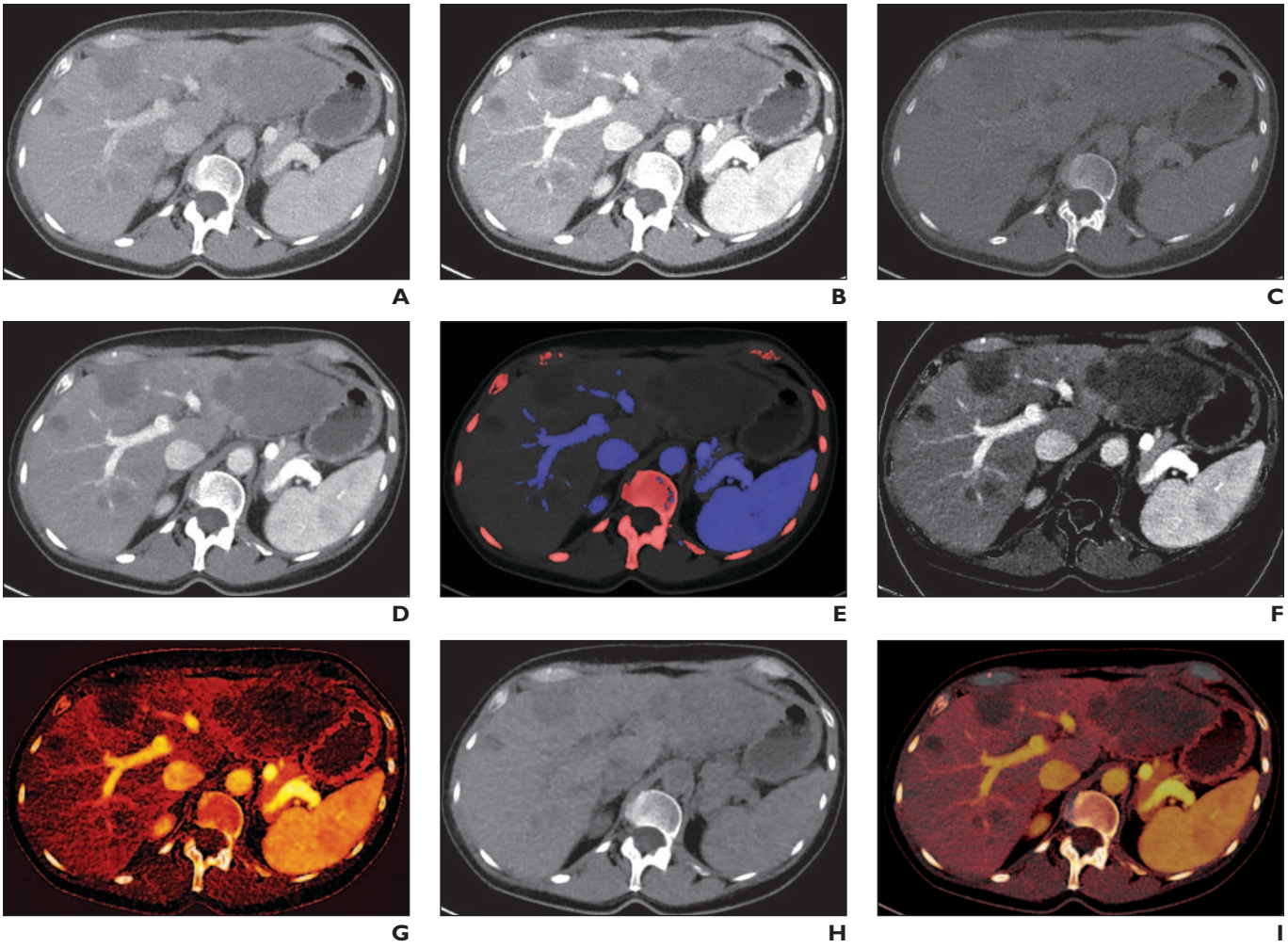


Fig. 4—Clinical example dataset obtained on dual-source CT scanner using 0.4-mm stannum filter at 140 kVp and 71 mAs and 100 kVp and 69 mAs with overall CT dose index of 5.7 mGy. Images were generated with Syngo dual-energy software (version VE32B, Siemens Healthcare) of 72-year-old woman with liver metastasis from colorectal cancer.
A, Image acquired at 140 kVp using stannum filter.
B, Image acquired at 100 kVp.
C, Quasi monoenergetic image extrapolated to 140 keV.
D, Optimum contrast image after "sigmoidal blending."
E, Algorithm differentiates iodine (*blue*) from calcium (*red*).
F, Angiographic image after bone removal.
G, Algorithm quantifies iodine by color-coding iodine in orange.
H, Virtual unenhanced image after iodine subtraction.
I, Fusion of color-coded iodine image and unenhanced image.

rapid kilovoltage-switching systems [5], about 70% for sequential acquisitions at different voltages (disregarding the diagnostic problem with the temporal offset), and up to 95% for quantum-counting detectors [13].

Besides the different technologic approaches to DECT, technologic strategies that allow dose reduction include tube current modulation, iterative reconstruction techniques, and new detector application-specific integrated circuits (ASICs) integrating photodiode and analog digital converters. These features offer special benefits for DECT because the CNR is improved in both half-dose acquisitions with the two energy spectra, so the gain in dose efficiency is even greater than in single-energy CT.

In brief, DECT does not necessarily imply an increased dose compared with single-energy examinations. However, effective patient dose and spectral contrast depend quite strongly on the technology used.

Postprocessing

There are generally two approaches to extract dual-energy information from projection data. A straightforward method is to subtract equivalent projections and apply filtered backprojection to reconstruct the difference as spectral information. Another way is to, first, reconstruct standard CT images consisting of voxels in Hounsfield units and then to use postprocessing algorithms to extract specific spectral information from the difference between the corresponding voxels. Currently, the more commonly used approach is the latter, with the image reconstruction system providing low- and high-kilovoltage images and a series of weighted average images. The average series integrates both acquisitions in a

low-noise image for immediate clinical evaluation. Dual-energy analysis is then performed on the dual-kilovoltage series using imaging-based algorithms. Meanwhile, 15 algorithms have been approved by the U.S. Food and Drug Administration for a variety of clinical applications, so the imaging-based approach is well established.

Three main types of algorithms are in use (Fig. 3). The first type optimizes images, the second type identifies or differentiates certain materials, and the third type quantifies a substance in the dataset. The output of the first algorithm consists of altered gray-level CT images, whereas the output of the latter two algorithms usually color-code substances—either several substances in different colors or the quantity of one substance on a color palette.

Examples of image optimization algorithms are monoenergetic images, in which the density (in Hounsfield units) for each voxel is extrapolated to a certain energy from the two density values at the acquired photon energies (monochromatic or monoenergetic [14] [Fig. 4C]), and nonlinear blending algorithms. Nonlinear blending algorithms combine high iodine contrast and low noise, which is referred to as "optimum contrast" [15] (Fig. 4D).

Differentiation algorithms define a slope between the density values at both acquired spectra and differentiate materials on the basis of the photoelectric effect within a certain density range—that is, colors are assigned on both sides of the slope (Fig. 3A). Examples for these algorithms include kidney stone differentiation (i.e., differentiation of uric acid from magnesium or calcium [16, 17]) or the differentiation of iodine and calcium (Fig. 4E). Another possibility is to eliminate certain substances from a dataset by identifying the substance and then filling in, for example, air density for the corresponding voxels (e.g., eliminate calcium for bone removal from angiographic datasets [18, 19]) (Fig. 4F).

Quantification algorithms use a three-material decomposition, quantifying one of three materials (Fig. 3B). A slope is defined by the density of two basic components (e.g., soft tissue and fat for the liver or soft tissue and air for the lung), and a second slope is defined by the photoelectric effect of the contrast material being quantified (i.e., iodine or xenon gas). The density values measured at both energies are then interpreted as a displacement from the first slope along the second one (i.e., as a certain iodine enhancement in an organ consisting of the two basic components). This enhancement is then color-coded or is also subtracted from the image (e.g., iodine in liver or lung parenchyma) [20–22] (Figs. 4G–4I).

Summary

DECT offers the possibility to exploit spectral information for diagnostic purposes. There are different technical approaches, all of which have inherent advantages and disadvantages especially regarding spectral contrast and dose efficiency. There are numerous clinical applications of DECT that are easily accessible with specific postprocessing algorithms.

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