Described by researchers as “one of the most exciting and promising developments in radiology in recent history,” dual-energy computed tomography (DECT) yields enhanced image contrast resolution by simultaneously or nearly simultaneously acquiring scan data at 2 different x-ray tube energy levels (typically 80 kV and 140 kV). X-ray attenuation differences between these energy spectra, measured by standard detectors, are then used to define, differentiate, or quantify the chemical composition of different tissues and materials. Because of this capacity for material differentiation and improved detection of iodine at low energies, DECT imaging and postprocessing can yield complex visualizations that are rich in structural and functional detail, allowing isolation and quantification of tissues and disease processes in better detail than usually is possible with traditional single-energy CT imaging—frequently without increased radiation doses to patients. Because DECT can visualize more diagnostic information than many other medical imaging modalities, it also can reduce the need for multiple imaging examinations and hence help minimize patients’ cumulative radiation doses.

DECT x-ray attenuation levels reflect interactions between photons at different energies and the atoms of scanned tissues and materials. The atomic number of key substances (e.g., uric acid, calcium, iodine, iron, and xenon gas) allow diagnostic images that differentiate and detail the internal chemical compositions of structural pathologies or disease processes. For example, DECT images can be used to differentiate kidney stones that contain calcium from kidney stones that contain uric acid, or to distinguish brain hemorrhages associated with other conditions.
with tumors from those with noncancer causes. This imaging modality also allows detailed characterization of nontissue materials and mitigates imaging artifacts associated with metal devices. In recently published preliminary studies, DECT also was able to differentiate heroin from cocaine, suggesting the possibility of sensitive identification of illicit substances in suspected smugglers’ gastrointestinal tracts, and potentially expanding imaging’s role in cross-border drug interdiction.

Although sequential or consecutive CT scan acquisitions at different x-ray tube voltages were first attempted in the mid-1970s, it was not until this century that sufficiently fast DECT scanners became widely available for clinical use, and the field continues to evolve. Routine visualization software now is standard at DECT workstations, allowing clinicians to address common questions in both acute (emergency) and chronic disease settings. For example, typical postprocessing options in DECT workstation software include automatic digital subtraction of bone from DECT angiography images, even in complex anatomic regions. Other DECT postprocessing options are monoenergetic imaging, lung perfusion imaging, and virtual noncontrast (hereafter, virtual unenhanced). 

DECT imaging now is used routinely for angiography and perfusion imaging, kidney stone characterization, reconstruction of virtual unenhanced images, quantification of iodine enhancement in solid organs, and visualization of uric acid crystals for diagnosis of gout. Emerging applications not yet in widespread clinical use include neuroradiology and oncology imaging for tumor detection, characterization, and monitoring during and after therapy. DECT colonography, ligament and tendon DECT imaging, neuroimaging, and lung ventilation imaging also are in development but are not yet in widespread clinical use.

History of DECT

Early in the development of CT technology, researchers recognized that knowing how target tissues respond to different x-ray energy spectra could reveal more information about that material’s chemical composition than could be discerned with single-energy CT scans. DECT was first described in the mid-1970s as a promising technique for visualizing the tissue signatures of healthy and diseased anatomies and originally involved sequential acquisition of 2 separate CT scans at different energy spectra. Although research confirmed that this approach could provide superior tissue characterization, early efforts to bring DECT into clinical practice were unsuccessful because of technological limitations. Early CT scanners frequently yielded very high levels of image noise at low energy spectra. In addition, long scan times and resulting delays between consecutive scans led to image quality challenges such as respiratory motion and anatomical registration (alignment) errors that resulted in artifacts. The use of contrast agents proved challenging because it was difficult to capture the same anatomy in the same position at the same phase of contrast enhancement. Also, the computing power available in the 1970s was an important limitation on how much postprocessing could be performed with large scan data sets. Furthermore, because 2 CT scans were required for each examination, scan-associated risks to patients increased; even if no registration, patient motion, or positioning errors occurred, exposure to ionizing radiation doubled. Repeat imaging after errors increased doses of both radiation and contrast agents even more.

It was not until the mid-2000s that a confluence of developments allowed the “rediscovery” of DECT. These developments included increased computing power; advances in CT technology such as slip-rings, volumetric (spiral), and multidetector CT (MDCT) scanners; and improved temporal resolution. Subsequently, researchers introduced scanners specifically designed to perform DECT scanning, and routine clinical DECT imaging arrived. Several DECT-capable scanner designs and models now are commercially available.

Despite researchers’ enthusiasm for DECT, concerns about radiation dose and health care cost control appear to have delayed its widespread clinical adoption, according to Schoepf and Colletti. “Acceptance and integration of these techniques into routine clinical algorithms have been slow and almost reluctant,” they wrote. The authors wondered why the academic community embraces DECT technology while clinical users seem indifferent. They suggested that “the semantic connotation of ‘dual-energy’ CT . . . may be associated
with higher (ie, double) radiation exposure compared with traditional single-energy CT techniques.\textsuperscript{7,8}

In reality, because DECT separates x-ray spectra into high- and low-energy components, the overall dose frequently should be comparable to doses associated with single-energy CT examinations, they noted.\textsuperscript{4} In addition, advances in postprocessing that improve DECT iodine detection and therefore image quality have allowed the reduction of iodine contrast doses by more than 50%.\textsuperscript{22}

**Contemporary DECT-Capable Scanner Designs**

While consecutive acquisition using single-energy helical CT scanners is possible, this approach suffers from the same problems that doomed efforts in the 1970s, although to a lesser extent. These problems include contrast enhancement timing and cardiac and respiratory motion artifacts caused by the delay between scans.\textsuperscript{2,4} Contemporary consecutive scanning involves algorithmic registration of the 2 data sets to correct for organ motion, but the contemporary clinical research literature discussed in this article is based on scanners specifically designed for DECT imaging.\textsuperscript{4} These DECT-capable scanners can perform the functions of a single-source CT scanner in addition to DECT scan acquisitions.\textsuperscript{22}

There are now several commercially available DECT scanners. These are either a single x-ray source with fast switching between 2 kilovoltage settings, such as GE Healthcare’s Gemstone Spectral Imaging mode Discovery CT750 HD scanner, or a 2-tube design, such as the dual-source CT (DSCT) SOMATOM Definition scanner manufactured by Siemens Medical Systems. Both types of DECT scanners depend on the simultaneous or near-simultaneous acquisition of data from 2 different x-ray energy spectra, measured by standard CT detectors (see **Figure 1**).\textsuperscript{2,3} A more recently developed alternative to these designs is the single-tube/dual-layer DECT scanner. Each of these approaches to scanner design has advantages and disadvantages. In addition, detector research and development efforts might yield literal photon-counting scanners in the future.

DECT generally refers to both DSCT and rapid kilovoltage-switching scanner designs, but some authors use DECT only for the latter. In this Directed Reading, DECT is used in the generic sense to refer to any CT scanner designed to acquire scan data using 2 photon

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**Figure 1.** Dual-energy computed tomography (DECT) scanner designs. Schematics show the difference between (A) dual-source computed tomography (DSCT) and (B) rapid-switching DECT scanners. In DSCT, separate x-ray tubes are used for high- and low-energy scan acquisitions. In rapid-switching DECT scanners, a single x-ray tube is rapidly alternated between high- and low-kilovoltage acquisitions. Reprinted with permission from Silva AC, Morse BG, Hara AK, Paden RG, Hongo N, Pavlicek W. Dual-energy (spectral) CT: applications in abdominal imaging. Radiographics. 2011;31(4):1033.
energy spectra, rather than only to rapid-switching scanners. DECT scanner technologies are evolving rapidly, and manufacturers’ documentation should be consulted for the specific parameters and protocols associated with a particular model of DECT scanner.

**Dual-Source CT Scanners**

DSCT scanners employ 2 independent x-ray tubes operating at different tube potentials, mounted orthogonally at roughly 90° to one another on the gantry, opposite their respective detectors (see Figure 1A). The high-energy tube operates at 120 kV or, more commonly, 140 kV; the low-energy tube operates at 80 kV or 100 kV. (Eighty kilovolts should not be used with obese patients.) Increasing the lower kilovoltage tube from 80 kV to 100 kV improves image quality for obese patients.

First-generation DSCT scanners had smaller detectors and limited fields of view (approximately 26-cm diameter). Subsequent DSCT models have 33-cm diameter fields of view, which is large enough to encompass the thorax and abdomen of most adults. However, 33-cm fields of view might sometimes be insufficient for fully analyzing target organs in obese patients.

DSCT scanners now also include integrated 0.1 mm tin filters to improve spectral separation. Each tube can be independently adjusted and filtered to fine-tune image quality and dose performance. Tin filters can be applied to remove low-energy quanta from the higher spectrum, improving image contrast.

Tin filtration of the high-kilovoltage DSCT tube improves discrimination of calcium and iodine without increasing radiation dose, compared with single-energy CT doses. The 2 x-ray sources operate simultaneously, resulting in cross-scatter radiation that contemporary scanners can detect, measure, and algorithmically correct (ie, remove) from acquired data sets. DSCT scanners tend to have good spectral separation between the high-kilovoltage and low-kilovoltage tube scans, and attenuation can be measured in Hounsfield units on virtual unenhanced or actually unenhanced images.

Today, most adult patients’ bodies can be scanned in a few seconds, minimizing the risk of registration errors and respiratory-motion artifacts.

**Single-Source Rapid Kilovoltage-Switching Scanners**

Rapid kilovoltage-switching DECT scanners contain only one x-ray tube and one detector; the lone tube’s voltage alternates rapidly between high and low energies (80 kV and 140 kV at intervals of 0.5 ms), acquiring both sets of data at each projection during a single gantry rotation (see Figure 1B). To accommodate the 2 scan acquisitions at each position, rapid kilovoltage-switching systems have a slightly slower gantry rotation time (350 ms vs 300 ms for DSCT, for example).

The x-ray tube output is at high energy (140 kV) approximately 65% of the time and low energy (80 kV) the other 35% of exposure time. Because the single-source DECT scanner design has only one tube and one detector, it offers good temporal and spatial registration, and equipment costs are lower than those of DSCT scanners.

The field of view for single-source DECT scanners is 50 cm—larger than is available with DSCT scanners. Single-tube fast kilovoltage-switching scanners offer good quantification of iodine density but have less well-separated high- and low-spectra (ie, greater spectral overlap) than DSCT scanners. Single-tube scanners also have higher noise for lower peak voltage and cannot visualize attenuation on virtual unenhanced images. Despite the short millisecond-level delays between acquisitions at different energy levels, rapid kilovoltage-switching systems yield “full interlaced” data sets because scan images are built using data from hundreds of acquisitions “during which the system switches back and forth.”

Detectors such as GE’s Gemstone Spectral Imaging for the Discovery CT 750HD scanner employ rare-earth–based oxides that replicate the chemical structure of garnet crystal and which are 100 times faster than semiconductor scintillator-based detectors such as those containing gadolinium sulfur dioxide (Gd$_2$O$_2$S).

With rapid kilovoltage-switching DECT scanners, the simulated Hounsfield unit values in monoenergetic iodine mapping images reflecting different extrapolated single-energy levels can vary considerably.
**Dual-Layer Detection Scanners**

Although not yet available commercially, dual-layer DECT scanner prototypes are now in operation.\(^1,^2,^3,^7\) Instead of generating different photon energy spectra, a dual-layer DECT system contains 2 layers of detectors, with sensitivities set to different photon energies emitted by a single x-ray tube operating at a single energy spectrum (kV).\(^1,^3,^7\) These detector layers allow the compartmentalization of photons into “energy bins.” The sensitivities are intrinsic to the scintillator components, such as a layer of Gd2O2S with an overlying layer of zinc selenide (ZnSe) or cesium iodide (CsI). The bottom detector layer is for acquisition of high-energy spectra, and the upper or overlying layer acts as the low-energy detector.\(^2\) These layered semiconductor scintillators have considerable spectral overlap, however. Dual-layer DECT scanners offer excellent temporal and spatial registration but limited spectral separation.\(^3,^4,^7\)

**Tissue and Material Properties and Attenuation**

Traditional single-energy CT imaging uses x-ray attenuation expressed as Hounsfield units that reflect relative tissue and material densities compared with water.\(^1\) This system limits the material differentiation that is possible in the human body, which consists of materials with overlapping CT values.\(^4\) In contrast, DECT exploits the interactions of photons and the materials with which they interact at different kilovoltage photon energies. This can provide much more anatomic and functional detail than single-energy CT scans.

X-ray attenuation varies with photon energy and the atomic qualities of target tissues or materials. Scanning a given target tissue or material at 80 kV yields different attenuations than scanning at 140 kV.\(^1,^6\) And scanning different tissues or materials at the same tube kilovoltage also yields different levels of x-ray attenuation. For instance, iodine exhibits peak attenuation at low energy, and bone calcium exhibits peak attenuation at high energies.\(^1,^6\)

DECT imaging depends on differences in both properties: x-ray attenuation at varying photon energies and attenuation differences among types of tissues and materials.\(^3\) X-ray attenuation depends importantly on 3 phenomena of interaction:

1. Photoelectric effect – low-energy photon and electron interaction, the so-called photoelectric effect, reflects the interaction of photons with atoms’ inner (K-shell) electrons; photons displace electrons from an atom, releasing energy as a photoelectron.\(^1,^6\) Photoelectric attenuation peaks for each element are known as those elements’ K-absorption edge (K-edge) values, which increase with increasing Z values (atomic number).\(^1,^6\) The K edge occurs when photon energy levels just exceed the K-shell binding energy, allowing the photon to displace an electron. The closer an x-ray tube’s energy level to the K-edge value of an element, the greater the attenuation.\(^1,^5,^6\)

Only materials composed of elements with relatively large Z value differences can be differentiated using DECT. Materials with high Z values are subject to greater photoelectric effect attenuation; the effect is weak for the most common elements that make up human tissues (hydrogen [Z = 1], carbon [Z = 6], nitrogen [Z = 7], and oxygen [Z = 8]). Conversely, calcium (Z = 20) has a stronger interaction with photons, and the strongest photoelectric effects occur with materials containing iodine (Z = 53), xenon (Z = 54), or barium (Z = 56), making these elements useful contrast agents.\(^2\) The K-edge values of calcium (4.0 keV; see Table 1) and iodine (33.2 keV) are higher than those of the hydrogen, carbon, nitrogen, and oxygen constituents of human tissues (≤ 0.53 keV), so they can be well differentiated in DECT images.\(^1,^6\)

2. Compton scatter – the loss of high photon energy to electrons in target material results in x-ray attenuation. Elements with low Z values are most affected by Compton scatter. This is the largest source of x-ray attenuation.\(^3\)

3. Rayleigh scatter – this is a weak contributor to x-ray attenuation. Rayleigh photon scatter occurs when photons encounter particles smaller than the photon’s wavelength.

Photoelectric effect and Compton scatter are key to understanding DECT imaging’s ability to differentiate and visualize key chemical components of tissues and disease processes.\(^1,^6\)
Energy Spectra

DECT requires production and detection of different x-ray energies, detector technology capable of differentiating attenuation of those energies, and distinct spectral properties in target tissues or materials.²,³ Peak tube kilovoltage is the measure of the higher limit of an x-ray tube’s photon energy output. The lowest and highest tube voltages typically are used for DECT acquisition, usually no lower than 80 kV and no higher than 140 kV.³,¹² Tube voltages below 80 kV are not useful because too much of the x-ray energy is absorbed by target tissues for differentiation.² Tube voltages higher than 140 kV are typically not available.²,³

Different elements’ K-edge values and the photoelectric effect’s energy-level dependence are central to tissue behavior during DECT scans. The K-edge value of iodine (33.2 keV) is closer to the low-energy DECT spectra (80 kV) than the high-energy 140 kV, for example, so its attenuation is higher in the low-energy data set.¹⁸

Radiation Dose

The ALARA principle (as low as reasonably achievable) is a central consideration for ensuring patient protection during any x-ray or CT examination. Because of the long-term radiation hazards posed by ionizing radiation, unnecessary CT examinations always should be avoided. This is particularly true for children, for whom the lifetime cancer risk posed by radiation is more pronounced because of the often decades-long latency period between irradiation and development of radiation-associated cancers. ALARA means that every patient dose of medical radiation should be the lowest possible to achieve specific clinical goals.

Unlike old sequential-scanning approaches to DECT with conventional CT scanners, contemporary DECT-capable scanners split x-ray spectra into high- and low-energy components so that the overall dose should be comparable to that of a traditional single-energy examination. In the 1970s, very early sequential-scan approaches to DECT resulted in at least a doubling of patients’ ionizing radiation doses.

The actual radiation doses involved in contemporary DECT imaging are a matter of contention and relatively little definitive study. Many authors report their single-institution assessments as universally applicable but without reporting controls used when comparing single-energy and DECT scanning protocols. Comparing reported radiation doses across studies is complicated by differences among research teams’ protocols, scan parameters (eg, rotation times and dwell times), anatomic regions studied, and the specific DECT-capable scanner equipment used.²²,²₈

Overall, however, the literature does not appear to support the common assumption that using contemporary DECT scanning doubles patients’ radiation doses in a manner similar to that of sequential scanning. Rather, the published evidence suggests that there is usually little, if any, “dose penalty” when DSCT scanners are used and that overall radiation doses associated with any type of DECT imaging fall well within national and international standards.¹,²² Most evidence for dose penalty came from studies of early-generation DSCT scanners. Newer scanners appear to be dose-neutral or dose-reduced compared with conventional single-tube CT scanners for comparable examinations.¹₂,¹₃

However, it is important to note that most radiation dose studies have employed DSCT rather than rapid-switching DECT scanners, and very little has been published regarding rapid-switching scanners’ radiation doses.¹₂,²₉,³⁰ Rotation time is an important variable in patient radiation dose; longer dwell times at 80 kVp “not

<table>
<thead>
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<th>Substance</th>
<th>K Edge (keV)</th>
<th>Atomic No. (Z)</th>
</tr>
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<tbody>
<tr>
<td>Hydrogen (H)</td>
<td>0.01</td>
<td>1</td>
</tr>
<tr>
<td>Carbon (C)</td>
<td>0.28</td>
<td>6</td>
</tr>
<tr>
<td>Nitrogen (N)</td>
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</tr>
<tr>
<td>Oxygen (O)</td>
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<td>8</td>
</tr>
<tr>
<td>Calcium (Ca)</td>
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</tr>
<tr>
<td>Iodine (I)</td>
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<td>Barium (Ba)</td>
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<td>56</td>
</tr>
<tr>
<td>Gadolinium (Ga)</td>
<td>50.20</td>
<td>64</td>
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</tbody>
</table>

Only lower the radiation dose but also optimize overall image noise,” Silva and colleagues noted. Rapid-switching scanners have longer dwell times for lower-energy scanning, which apparently puts their radiation doses generally within those published by the American College of Radiology for the volumetric abdominal CT dose index (CTDIvol) of 25 mGy. Nevertheless, rapid-switching scanners are by some assessments associated with possible elevations in radiation dose of up to 20% or even 40% compared with conventional single-tube CT scans of the same patients.25,30

It also is important to keep in mind that DECT offers “substantial incremental diagnostic gain . . . which often obviates multiple phase CT or additional clinical tests that involve radiation, thus reducing the net exposure of the individual patient.”22 In other words, DECT typically yields more clinically relevant information than a traditional single-energy CT examination, frequently at comparable radiation doses, although this is most clear for DSCT scanners.29 Radiation dose comparison studies of different imaging modalities and techniques frequently ignore signal-to-noise ratios and image quality and thus the diagnostic usefulness of the resulting images.7 When dose comparisons are made between different scanners for images of comparable quality, there appears to be little basis for the assumption that DECT involves increased radiation doses.9 Indeed, DECT images using blended high- and low-energy data might offer improved contrast-enhanced contrast-to-noise ratios to single-tube 120 kV images without an increase in radiation dose to the patient.1

Despite similar concerns specific to cardiac DECT radiation doses to the heart, the available evidence suggests that coronary CT angiography performed with first-generation DSCT and rapid-switching DECT scanners yields lower radiation doses than does coronary angiography performed with a single-source 16-slice MDCT scanner.1,31 Kerl et al reported no significant differences in diagnostic image quality for DSCT and rapid-switching DECT scans.31 They noted lower angiographic radiation doses and comparable quality for images obtained using rapid switching compared with DSCT scans.31 Similarly, it appears that pulmonary angiography using either second-generation DSCT or rapid-switching DECT involves radiation doses similar to 64-slice MDCT angiography and lower doses than those associated with conventional single-energy 128-slice MDCT angiography.1,2,32,33

Finally, virtual unenhanced imaging represents an important advance in terms of radiation dose management. With this technique, a single contrast-enhanced DECT scan acquisition provides both precontrast and contrast-enhanced images that would require 2 separate scan acquisitions with conventional single-energy CT. Replacing unenhanced conventional CT scans with second-generation DSCT virtual unenhanced CT images can cut radiation doses by 50%, according to several studies of kidney stone or kidney mass imaging.1,28,34 Virtual unenhanced CT images might be a valuable way to reduce patients’ cumulative radiation doses when monitoring or follow up requires frequent reimaging, such as monitoring aortic stent grafts, which typically are scanned every 6 months.1

Image Visualization

DECT images can be used to produce standard CT images like those obtained using conventional single-energy CT, or information-rich, material-specific images not available using conventional CT imaging.24 DECT data sets can be used to depict anatomies in either 2 or 3 dimensions, often using similar projections, reformations, and renderings as those used in conventional CT visualizations, such as 2-D maximum-intensity projection (MIP) or curved planar MIP. These projections employ ray-casting algorithms to capture information from underlying volumetric data sets; the path of a linear ray of imaginary light can be calculated through data representing a scanned volume, logging attenuation values encountered at sampling points along its path. The brightness of pixels in the resulting 2-D projections reflect the underlying attenuation values encountered by the cast ray at all sampling points.35,36

Postprocessing allows rapid, dynamic review of DECT data.21 The postprocessing algorithms used with different DECT-capable scanners are proprietary software but can be described in general terms. Postprocessing algorithms can be used to generate virtual unenhanced or enhanced images, digital subtraction images, color-coded tissue-specific visualizations, and quantification images reflecting the densities
of a specific element. Postprocessing DECT scan-
acquisition data from one or both energy-spectra data
sets allows considerable diagnostic-imaging flexibility,
including extrapolated monoenergetic images, iodine
mapping of contrast distributions and densities, digital
elimination of anatomies or iodine contrast-affected
anatomies at threshold densities, and algorithmic
extrapolation of data visualization to emulate higher-
energy scans.5,14 (DSCT visualizes iodine density as
average enhancement in Hounsfield units, whereas rapid-
switching DECT quantifies iodine distributions in mil-
ligrams per milliliter concentrations.15) This flexibility
and the detailed information on the resulting images
might frequently allow clinicians to forgo additional
CT or scintigraphy imaging, further reducing a patient’s
cumulative imaging-related radiation dose.5,14

Once CT scan data have been acquired, 3 algo-
metric steps are performed to calculate clinically useful
images from the data:

- Registration – anatomically aligns data from dif-
  ferent energy spectra.
- Visualization – involves the reconstruction and
  postprocessing of registered data to generate cli-
  nically useful images.
- Segmentation – is a type of postprocessing that
  involves partitioning scan data into specific ana-
  tomic features, distinguishing anatomic boundar-
  ies or surfaces of interest from adjacent anatomies
  to allow visualization or digital subtraction of
  specified surface contours or internal components.
  Segmentation increasingly relies on automated
  workstation software with little input from the
  technologist.36 With DECT imaging, segmenta-
  tion is largely a matter of material decomposition.

DECT visualization terminology differs somewhat
among authors, with algorithms and image types being
variously classified as material selective or material spe-
cific (decomposition images, mapping atomic number
and density), nonmaterial specific or energy specific (also
called energy selective, reflecting attenuations at a partic-
ular photon-energy level), for example.4,14 Alternatively,
algorithms can be described simply as differentiating or
quantifying.2

Raw, or projection, data are simply the x-ray values
for each voxel (3-D pixel) of imaged anatomy, compared
with reference water values. Initially, for rapid-switching
(single-source) DECT-acquired data, projection images
typically are reconstructed automatically using data from
a specified energy level, such as 140 kVp.24 Alternatively,
weighted averaging or blending reconstructions can yield
projection images on a standard workstation screen dis-
play matrix of 512 pixels by 512 pixels.24,30 Blending low-
and high-energy data yields higher-quality images than
when only one data set is used, partly because the result-
ing images benefit from the high contrast of low-energy
acquisitions and the low noise of high-energy acquisi-
tions.4 Data from the 2 energy spectra are blended,
using one or the other energy level (80 kVp or 140 kVp)
for each imaged voxel.39 Weighted blending algorithms
employ high- and low-energy data blending ratios, typi-
cally set at 0.3, which weights 70% of the high-energy
scan data and 30% from the low-energy data set.4

These optimization imaging techniques can yield
monochromatic gray-scale CT images for routine cli-
nical interpretation, using weighted average images from
high- and low-energy scan acquisition data sets and
making use of all acquired data. The resulting images
are similar to images obtained with single-source CT
scans at 120 kV but are less affected by beam-hardening
artifacts and have more CT value accuracy than con-
ventional single-energy CT imaging.5,14,24 Because
DECT data sets can yield monochromatic images at
multiple energy values, image sets or series can be
produced that allow comparison of optimal contrast
visualizations of adjacent anatomies.24 More advanced
nonlinear optimization postprocessing algorithms
allow for more mathematically sophisticated blending
of high- and low-energy scan data, yielding superior
contrast-to-noise ratios.2,5,6

Material-Specific Image Processing

The nonmaterial-specific visualizations described
above yield standardized diagnostic images of sufficient
detail for routine review and interpretation by radiolo-
gists without extensive image postprocessing.4 These
images are essentially structural, similar to conven-
tional single-energy CT imaging. However, more refined
and informative DECT images can be produced using post-
processing software to reflect the additional information
contained in DECT scan data sets.
Material-specific processing techniques also are referred to as material density, material differentiation, or material decomposition. Material-specific images visualize attenuation differences between materials in low- and high-energy spectra data sets, and then either brighten or color-code each pixel accordingly or subtract it from the image. For example, automatic digital removal of bone calcium or other material can be achieved using differentiation postprocessing to reveal unobstructed views of blood vessels or other target tissues. Differentiation algorithms also compare and visually depict, through color coding, the photoelectric effects at different energy levels for tissues or materials with different K-edge values. For instance, these algorithms allow visual differentiation of fat from calcium and iodine, or uric acid kidney stones from stones composed of calcium.

Contrast-enhanced DECT images are generated using scan data acquired after intravenous injection of iodinated contrast materials. Iodine can be subtracted using material decomposition algorithms, creating so-called virtual unenhanced CT images that are similar to unenhanced or precontrast images obtained with conventional single-energy CT scanning. Image interpretation studies have shown that virtual unenhanced DECT images have similar image quality as unenhanced conventional CT images. This means that a single contrast-enhanced DECT acquisition generates scan data equivalent to 2 conventional CT perfusion scans (ie, precontrast and contrast), thereby reducing patients’ radiation doses. Because these images are generated from the same data set as the contrast-enhanced visualizations, they do not suffer from respiratory or organ motion or positioning mismatch artifacts that can affect traditional sequential-acquisition contrast CT scans.

Quantification algorithms visualize the physical densities of substances within a given anatomic region, such as arterial plaques’ calcium components or, in theory, iron stores in liver tissue. Paired iodine and water densities for a region of interest obtained using contrast-enhanced DECT scanning can be depicted either in terms of milligrams per milliliter or as visual enhancement, with the density of iodine in a given region of interest depicted as varying pixel brightness values.

Clinical Applications
Since their development less than a decade ago, DECT-capable scanners have become routinely used at university hospitals to aid the clinical assessment of various acute (emergency) conditions and chronic diseases. These include identifying and characterizing kidney stones, gout, and pulmonary embolism (PE), as well as reducing metal-associated imaging artifacts. The speed and detailed information of contemporary DECT images make them a particularly valuable diagnostic tool in emergency settings when PE is suspected. In addition, many other applications are in development, including kidney and liver lesion and tumor detection, lung ventilation imaging, heart and pulmonary hyper-tension imaging, bone tumor imaging, and detection of ligament and tendon tears in knee trauma.

From the patient’s perspective, DECT and single-energy scanning procedures are difficult to distinguish. They require the same patient preparation and positioning, and with modern spiral CT technology, both single-energy and DECT examinations are quick procedures for patients. The patient is positioned in the center of the scanner because the field of view of the second tube and detector is smaller than conventional arrays. Manufacturers’ documentation and institutional operating procedures should be consulted for protocols specific to the scanner models used in a particular radiology department.

Thoracic Imaging
Thoracic DECT examinations typically involve intravenous injections of 50 mL to 80 mL of 350 mg/mL-concentration iodinated contrast agent. Because DECT perfusion imaging’s improved iodine contrast visualization, smaller volumes (eg, 9.6 mL) of iodinated contrast agent can be employed. This represents an important advance for diagnostic imaging of patients with renal failure.

Pulmonary DECT Imaging: Perfusion and Ventilation
Because DECT better differentiates and characterizes lung tissues than traditional single-source CT scans and because it visualizes both anatomic and functional information, it is well suited for pulmonary imaging in both acute and chronic disease settings. For example,
DECT allows excellent contrast-to-noise ratios for identifying life-threatening PE with excellent vascular enhancement, and workstation image extrapolations at different energy levels (so-called “spectral surfing”) between 90 keV and 130 keV allow image optimization for visualization of clots. Important DECT applications for lung imaging include iodine contrast-enhanced perfusion scans for PE assessment and, potentially, diagnostic imaging of suspected pulmonary hypertension, chronic obstructive pulmonary disease (COPD), emphysema, and lung cancer. Iodinated contrast DECT imaging also shows promise in posttreatment surveillance for tumor recurrence. Contrast-phase DECT iodine maps visualize the distribution of lung tissue perfusion and regions with perfusion defects associated with disease or injury.

Iodine-concentration images depict perfusion of the lung tissues similar to the information yielded in traditional perfusion scintigraphy. Currently, the primary application for DECT perfusion imaging is PE assessment. PE is a potentially life-threatening blockage within the pulmonary vasculature. It is a frequent complication of deep vein thrombosis in the legs. Thromboemboli typically arise in leg or pelvic veins and can break free from the vascular wall. They then travel to the heart and subsequently into the lungs, where they can cause life-threatening blood stream blockages.

The lungs are anatomically complex organs with correspondingly complex vasculature. The right and left pulmonary arterial branches bifurcate repeatedly, creating a vascular “tree” or rootlike system of descending vessels with increasingly smaller diameters. Clots can travel downstream until they lodge in a smaller-diameter vessel, preventing downstream tissues from receiving sufficient nutrients and oxygen. In severe cases, where blood vessel obstruction is widespread and chronic, right heart function becomes imperiled by the resulting hypertension. Together, DECT visualization of intravascular thrombi and arterial obstruction, lung perfusion, and right heart function can identify chronic PE.

The perfusion scan data sets yield both functional and anatomic or morphological information that can be used to assess PE. The use of iodine-based contrast agents has allowed improved DECT detection of PE through clot and perfusion defect visualization, which together indicate occlusive embolism. Pulmonary nodules also can be characterized using DECT with iodinated contrast perfusion imaging. PE-associated perfusion defects typically are found in the lungs’ peripheries and appear as wedge-shaped triangular hypodensities (see Figure 2). In contrast,
beam hardening or lung-based motion artifacts that can emulate pulmonary perfusion defects in patients who do not have PE tend to appear crescent shaped.44

Iodine subtraction postprocessing of the perfusion scan data sets yields virtual unenhanced images that can help identify these PE intravascular blockages responsible for perfusion defects.14

Lung perfusion scans map iodine content throughout the lungs, comparing high- and low-energy scan acquisition data for material differentiation.15 A lung perfusion scan protocol intended to visualize pulmonary arterial anatomy and perfusion in a single scan using the Siemens Definition Flash spiral DECT scanner is depicted in Table 2.

It is important to note that obese patients’ peripheral lungs might not fall within the field of view of the second smaller tube and detector in a 2-source DECT scanner.43

The timing of scan acquisitions in relation to contrast injection is a crucial factor in image quality.45 Automatic power injectors deliver a contrast agent at a flow rate of 3.5 mL/s with a 50 mL saline flush.22 Automated contrast media bolus-triggered scan acquisitions are available with most contemporary DECT-capable scanners. These systems track attenuation of a target artery and trigger scanning when iodine concentration in that vessel reaches a predetermined threshold.43,45 For DECT perfusion imaging, high-concentration contrast material is injected with the patient positioned for a caudocranial scan to avoid contrast-associated image artifacts, and bolus tracking triggers scan acquisition when iodine reaches the pulmonary artery.43 The postinjection scan delay should be 7 seconds. This allows sufficient time for the contrast to enter the distal lungs.43

Severe cases of emphysema are sometimes treated with lung volume reduction surgery.39

### Table 2

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Definition</th>
<th>Definition Flash</th>
</tr>
</thead>
<tbody>
<tr>
<td>Scan mode</td>
<td>Spiral dual energy</td>
<td>Spiral dual energy</td>
</tr>
<tr>
<td>Scan area</td>
<td>Diaphragm to lung apex</td>
<td>Diaphragm to lung apex</td>
</tr>
<tr>
<td>Scan direction</td>
<td>Caudocranial</td>
<td>Caudocranial</td>
</tr>
<tr>
<td>Scan time (s) for 300-mm length</td>
<td>11</td>
<td>9</td>
</tr>
<tr>
<td>Tube voltage A/B (kV)</td>
<td>140/80</td>
<td>100/140 Sn (tin filter)</td>
</tr>
<tr>
<td>Tube current A/B (quality ref mAs)</td>
<td>47/235</td>
<td>89/76</td>
</tr>
<tr>
<td>Dose modulation</td>
<td>CARE dose 4-D set “on”</td>
<td>CARE dose 4-D set “on”</td>
</tr>
<tr>
<td>CTDIvol (mGy)</td>
<td>9.2</td>
<td>7.3</td>
</tr>
<tr>
<td>Rotation time (s)</td>
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<td>0.28</td>
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<tr>
<td>Pitch</td>
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<tr>
<td>Slice collimation (mm)</td>
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<td>0.6</td>
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<tr>
<td>Acquisition</td>
<td>14 × 1.2 mm</td>
<td>64 × 0.6 mm</td>
</tr>
<tr>
<td>Slice width (mm)</td>
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<td>2</td>
</tr>
<tr>
<td>Reconstruction increment (mm)</td>
<td>1.4</td>
<td>1.4</td>
</tr>
<tr>
<td>Reconstruction kernel</td>
<td>D30f</td>
<td>D30f</td>
</tr>
</tbody>
</table>

assessment involves regional perfusion assessment, which can be achieved using virtual unenhanced and iodine-mapping DECT images.\(^9\)

Once pulmonary perfusion scan data are acquired, postprocessing can require a dedicated DECT workstation (or server) for up to 10 minutes.\(^{22}\) “Spectral surfing” reconstructions at various extrapolated energies between 40 keV and 80 keV allow image-to-noise optimization.\(^{22}\)

Three postprocessing techniques for iodinated-contrast DECT lung perfusion images commonly are used.\(^{15}\) Native perfusion scans yield gray-scale or color-coded visualizations of blood supplies throughout the lungs (see Figure 3). MIPs or curved-planar MIP scans similarly visualize blood supplies as gray-scale or color-coded images. Fused scans overlay MIP and native perfusion scans.

In addition to blood perfusion, regional patterns of air ventilation of the lungs can be altered by lung diseases such as infarction, COPD, and cancer.\(^{23}\) Traditional nuclear medicine lung perfusion/ventilation imaging (so-called VQ scintigraphy) is widely used to assess both blood flow and oxygen ventilation patterns in the lungs—that is, how well blood circulates through the lungs and how well atmospheric air reaches all regions of the lungs.\(^{43}\) This can help to pinpoint PE foci, as well as to detect and characterize diffuse lung diseases such as pneumonia or COPD. DECT is emerging as superior to single-photon emission CT (SPECT) for VQ imaging.\(^{29}\)

Inhaled-contrast DECT ventilation imaging using noble gases with high Z values, such as xenon and krypton, is under investigation for high-resolution visualizations.\(^{22}\) In particular, xenon-enhanced DECT ventilation scans that yield selective low- and high-energy axial, coronal, and sagittal 2-D images show promise.\(^{23,39}\) (MIP and weighted blending images with 30% low-energy data and 70% high-energy data also are used to create morphologic reconstructions for comparison purposes.\(^{23}\)) These images also can be volume rendered in postprocessing (see Figure 4).\(^7\)

Xenon is a stable, radiopaque noble gas.\(^{23}\) Its Z value (54) is similar to iodine’s (53), making xenon an excellent contrast agent for radiographic imaging.\(^{23}\) Adverse reactions associated with xenon have been described as “mostly mild and uncommon” but include respiratory rate delay longer than 10 seconds in 3.6% of patients who inhale 32% xenon gas.\(^{23}\) Other adverse events occur less frequently, in fewer than 1% of patients, and are transient, resolving after xenon inhalation is halted. These include seizures, headache, nausea, vomiting, and altered neurological status.\(^{23}\) Patient preparation involves tightly fitting patients with a positive-pressure face mask for xenon delivery. The patient begins by inhaling air with at least 60% oxygen without enriched xenon for 2 to 3 minutes, then inhales xenon-enriched gas (30% xenon/70% oxygen).\(^{23}\) Patient respiration rate, inhaled xenon-gas concentration, blood oxygen saturation, and blood pressure are monitored before and during the xenon-enhanced DECT ventilation examination.\(^{23}\) After the examination is complete, respiration

---

**Figure 3.** Pulmonary embolism DECT imaging of a woman with chest pain and bilateral multiple thromboemboli. Coronal DECT images (A, B) and sagittal (C) pulmonary perfusion DECT images show decreased perfusion (dark blue areas) in the right upper and lower lobes and left lobe, compared with areas with normal perfusion (light blue areas). Curved planar maximum-intensity projection (MIP) images visualize emboli in right (D) and left (E) inferior pulmonary arteries. Reprinted with permission from Aran S, Besheli LD, Karcaaltincaba M, Gupta R, Flores EJ, Abujudeh HH. Applications of dual-energy CT in emergency radiology. AJR Am J Roentgenol. 2014;202(4):W315.
rate, blood oxygen, and blood pressure should continue to be monitored for several minutes. Protocols vary but include scan acquisitions during xenon wash-in (60 to 90 seconds) and washout phases (120 seconds), with CT acquisition every 20 to 30 seconds. Alternatively, static, reduced-dose whole-lung ventilation DECT imaging can be undertaken; after 2 to 3 minutes of high-concentration oxygen inhalation, the scan is acquired once during the end of the wash-in phase (at approximately 60 to 90 seconds, when xenon concentration in the lungs peaks at ~30%) and at the end of the washout phase (at approximately 2 to 3 minutes, when xenon concentration has returned to near zero). Xenon images can suffer from gravity artifacts, with higher concentrations of xenon accumulating in the ventral lungs when the patient is supine. Although xenon-enhanced DECT ventilation imaging is still largely investigational, this modality shows promise in the diagnostic imaging of COPD, asthma, bronchiolitis, and bronchial atresia, in addition to PE. Research is underway to employ xenon-enhanced DECT for lung surgery planning, assessing residual postsurgical lung function, and mapping regional ventilation defects associated with diffuse lung diseases such as COPD.

Cardiovascular Imaging

Coronary artery disease (also known as coronary heart disease) is a reduction of the heart’s blood and oxygen supplies caused by intraluminal narrowing of the small coronary arteries by atherosclerotic plaques. DECT can help confirm coronary artery disease by visualizing coronary arterial stenosis, blood flow dynamics, and right heart function and myocardial perfusion, in a single examination. Detecting intra-arterial plaques, characterizing their distribution and morphologies, and differentiating their compositions can help in treatment planning and determining patient prognosis. Conventional single-tube CT scanners cannot well differentiate materials present in different types of atherosclerotic plaques, limiting this modality’s utility for plaque imaging. DECT scans, conversely, can differentiate heavily calcified plaques from other atherosclerosis. Much more research is required before DECT becomes routine for diagnostic imaging of plaque. In addition, because of the iron content of blood, iron quantification also should be able to detect bleeding into atherosclerotic plaques, a risk factor for plaque instability and clot formation.

While DECT angiography is technically capable of assessing myocardial perfusion and right heart function, it has not yet been widely adopted for these applications. Myocardial perfusion defects can be accurately detected using iodine-enhanced DECT, potentially providing an alternative to other, more commonly employed imaging modalities, including SPECT and magnetic resonance (MR) imaging. Whether DECT can detect myocardial infarction and ischemia as well as MR is debatable. Published data indicate up to 92% sensitivity and 93% specificity for DECT in detecting myocardial perfusion defects. However, a 2013 analysis found that although late iodine enhancement data in cardiac DECT compare well with MR imaging for detecting chronic myocardial infarction, DECT iodine mapping is less accurate than MR imaging. The authors of that study found that postprocessing with blending algorithms improves scar volume estimation in myocardial infarction DECT.

imaging, however.48 Echocardiogram-gated DECT scanners are now commercially available and can further reduce imaging artifacts.22

Aortic DECT iodinated contrast-enhanced angiography promises to monitor aortic endovascular aneurysm repairs for endograft leaks and at lower radiation doses than those possible with repeated single-source MDCT scans.22,49 Virtual unenhanced DECT images have been deemed a sufficiently “reasonable approximation” of true noncontrast CT images to replace the latter.49 However, large clinical studies have not been undertaken to confirm preliminary support for this indication for DECT imaging. DECT imaging of other aortic disorders, such as intramural hematoma, also has not been well studied.49

Abdominal Imaging

Kidney Stones

The chemical composition of kidney stones can guide clinical management decisions. For example, surgical removal is indicated when stones are not composed of uric acid, such as stones containing cysteine or calcium oxalate monohydrate. Uric acid kidney stones, conversely, can sometimes be treated noninvasively through alkalization and dilution of urine, depending on stone diameter.4 Conventional single-energy CT does not appear to reliably reveal the chemical composition of kidney stones, but DECT can do so with postprocessing software that calculates the ratio of attenuation between the high- and low-energy data sets.4,25,50 In addition to differentiating uric acid stones from stones with other chemical compositions, DECT also can differentiate various types of nonuric acid stones, including cysteine, calcium oxalate, calcium phosphate, and apatite stones (see Figure 5).4,31,52

Stones in the kidneys and ureters have been identified with iodinated contrast subtraction postprocessing of DECT image data sets that yield virtual unenhanced images. In addition, this approach identifies stones with less ionizing radiation to patients than would result from conventional sequential CT acquisitions.4,53 However, authors of one study cautioned that only stones larger than 2.9 mm in diameter were reliably detected using iodine-subtraction virtual unenhanced DECT images because smaller and low-attenuation stones are frequently inadvertently deleted from these images.53 While virtual unenhanced DECT urography appears to be useful for larger stones, which are more likely to be symptomatic, DECT should not be relied on to exclude the possibility of kidney stones smaller than 3.0 mm in diameter.53

Iodine contrast-subtracted virtual unenhanced DECT images also can help differentiate between types of renal lesions, such as hemorrhagic cysts and solid tumors.4 Hemorrhagic kidney lesions are visible on both contrast-enhanced and virtual unenhanced DECT images. Iodinated contrast-enhanced images allow color-coded quantification of iodine within a lesion.9

Figure 5. DECT characterization of kidney stones. A. DECT axial image shows left kidney calculus (stone) with attenuation of 825 HU at 140 kVp. B. This image has 1250 HU attenuation at 80 kVp. C. The stone was determined to be composed of calcium oxalate monohydrate using DECT postprocessing algorithms. Reprinted with permission from Coursey CA, Nelson RC, Boll DT, et al. Dual-energy multidetector CT. Radiographics. 2010;30(4):1045.
Liver

Iodinated contrast-enhanced DECT reveals focal liver lesions such as calcified or well-vascularized metastatic tumors and, in theory, also should be able to visualize diffuse conditions such as hemochromatosis (iron overload, which can cause cirrhosis) and fatty liver. Fatty liver is seen in association with obesity, diabetes, chronic alcoholism, hyperlipidemia, and anorexia. Fat deposits in the liver can mimic tumors, making material differentiation important. DECT’s ability to differentiate fatty liver from other tissues appears to be compromised when hemochromatosis also is present.

The evidence base for DECT iron quantification of liver tissue is limited. While specificity is nearly perfect, sensitivity is so poor that some authors have recommended it not be used for this application, especially at low levels of iron overload.

For other DECT liver imaging applications, such as identifying focal lesions, tube voltage is set to 140 kV and 80 kV, with the patient in a supine position. Contrast agent protocol is the same as for single-tube CT liver imaging, with contrast agent and saline injected with a power injector through an indwelling venous catheter at a flow rate of 2.5 mL/s to 5.0 mL/s. The resulting data sets are postprocessed using blending at a ratio of 70% 140 kV data and 30% 80 kV data.

Surprisingly, research is limited regarding the promise of virtual unenhanced DECT for reduced radiation dose imaging of the liver. However, a scan protocol for the Siemens Definition Flash spiral DECT-capable scanner is as follows:

- Scan area – diaphragm to iliac crest.
- Scan direction – craniocaudal.
- Kilovoltage – 100 kV/140 kV (with tin filter).
- Milliamperes seconds – 230 mAs/177.7 mAs.
- Dose modulation – CARE dose 4-D set “on.”
- CTDI_vol – 18.2 mGy.
- Rotation time – 0.50 seconds.
- Pitch – 0.60.
- Slice collimation – 0.6 mm.
- Acquisition – 32 mm × 0.6 mm.

Musculoskeletal Imaging

DECT plays an emerging role in several important musculoskeletal imaging applications, including detection of collagen, ligament, and tendon damage in the knees, wrists, and hands, and imaging of tissues adjacent to or near metal medical implants. Early-stage research suggests that bone marrow edema (water accumulation, swelling, and bruising) is accurately imaged with DECT. As the baby boom generation ages, metal implants will become increasingly common among patients seen at medical imaging departments. Metal implants such as hip replacements and vertebral fusion implants can cause radiographic artifacts such as beam hardening and streaking, which can complicate imaging-based assessment and monitoring for fractures, infections, and adjacent soft-tissue inflammation. Overcoming metal-associated imaging artifacts has proven to be challenging for conventional single-source CT scanners (see Figure 6.)

Colonoscopy screening for early detection of colon cancer is recommended for patients older than 50 years. Conventional single-source CT colonography screening for polyp and tumor detection typically involves fecal tagging with contrast agent for digital subtraction, which facilitates differentiation of fecal matter from polyps and tumors in the resulting images. DECT can perform the same examination and also can be used with iodine mapping and virtual unenhanced images to achieve the same goal of differentiating stool from potential cancer or precancerous tissue. These latter methods of differentiating stool from polyps might be particularly useful for patients who cannot undergo invasive optical colonoscopy or laxative and fecal-tagging preparation before imaging. Indeed, some researchers have reported preliminary feasibility-study evidence that DECT colonography might enable patients to forgo laxatives and fecal-tagging altogether, effectively visualizing colorectal tumors and even potentially differentiating tumors from nonmalignant polyps and stool.
Beam hardening is one problem with imaging around metal implants; low-energy photons are absorbed by metal alloys, yielding dark streaks in target tissue behind or adjacent to the metal object. \(^5\) Weighted blending of high- and low-energy DECT acquisition data sets can reduce the incidence and magnitude of these artifacts. \(^6\) DECT workstations even allow the operator to review artifact effects by changing the weighted energy levels to identify values that minimize artifacts without losing soft-tissue resolution. \(^7\) This optimal “sweet spot” typically is in the 105 keV to 133 keV range. \(^8\)

Because of superior anatomic registration between high- and low-energy data sets in DECT, it has been possible to develop software that compares data sets and corrects for beam hardening, streaking, and blooming artifacts that are sometimes seen at lower-energy acquisitions of anatomy near metal objects such as hip-replacement alloy prostheses. \(^4\)

**Gout**

Gout is a metabolic disorder that leads to the accumulation of monosodium urate monohydrate (uric acid) crystals within joints, triggering periodic and profoundly painful local inflammation reactions. \(^5\) Gout can be truly debilitating, preventing normal walking and other activity. Acute episodes have been described as “one of the most painful experiences reported throughout medical history.” \(^59,60\) For reasons that remain unclear, its prevalence appears to be increasing and currently affects more than 8 million Americans. \(^59\)

Gout frequently is diagnosed clinically, based on symptoms alone, but definitive diagnosis involves microscopic analysis of joint fluid, obtained in an aspiration procedure that carries a slight risk of septic arthritis. \(^58\) Before DECT imaging for suspected gout, other possible diagnoses are ruled out, including rheumatoid and other forms of arthritis, osteomyelitis, and trauma. \(^58\)

Some research suggests ultrasonography can reliably and sensitively detect gout, but generally speaking, non-DECT imaging modalities are not widely accepted as reliable for gout detection. DECT can sensitively identify this disorder and has therefore been described as a revolution in gout diagnosis and treatment (see Figure 7). \(^58\)

For suspected gout of the upper extremities, patients are positioned craniocaudally (head-first) in a prone position (face-down), with arms forward of the head, palms down. \(^58\) Radiolucent sponges and pillow may be used to support palms and digits, and head, respectively. \(^58\) For ankles, patients are positioned feet-first and

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**Figure 6.** Reduction of metal implant–associated CT artifacts in DECT. A. Coronal conventional CT image obtained after bilateral total hip replacement arthroplasty showing beam hardening artifact that precludes accurate evaluation of intrapelvic soft tissues. B. DECT image of the same anatomy shows substantial reduction of streak and photon starvation artifacts, allowing clear depiction of anatomic structures adjacent to the metal prostheses and intrapelvic soft tissues. Reprinted with permission from Silva AC, Morse BG, Hara AK, Paden RG, Hongo N, Pavlicek W. Dual-energy (spectral) CT: applications in abdominal imaging. Radiographics. 2011;31(4):1041.
supine (face-up) with bent knees. Imaging of the left and right hands, wrists, elbows, knees, ankles and feet should be undertaken in a DECT gout examination.

A published DECT protocol for gout imaging is as follows:

- **mAs (kV 140)** – 100 mAs for upper extremities, 120 mAs for lower extremities.
- **mAs (kV 80)** – 425 mAs for upper extremities, 400 mAs for lower extremities.
- **Pitch** – 0.55 for both upper and lower extremity imaging.
- **Collimation** – 64 mm × 0.6 mm for both upper and lower extremities.

Uric acid and calcium material decomposition algorithms for gout detection and visualization are available with contemporary DECT-capable scanner workstations and servers. Image artifacts around nail beds on the fingers and toes and at tendons on the feet and hands can sometimes be mistaken for gout. Metallic objects also can create artifacts that can be misinterpreted as gout on DECT gout examinations.

**Oncology Imaging**

Early, accurate tumor detection, staging of tumors’ extent, vascularization, local tissue invasion, distant metastatic spread to other organs, and early radiographic assessment of treatment responses are longstanding goals for cancer imaging. Although DECT is not yet in widespread, routine clinical use for cancer imaging, promising oncology applications for this modality include tumor detection, characterization, and treatment planning and monitoring. Reduced cumulative radiation dose is a potential advantage of DECT compared with traditional CT cancer imaging because of the frequent and repeated imaging typically undertaken in diagnosing, characterizing, treating, and monitoring malignancies.

DECT allows rapid anatomic and functional whole-body cancer imaging and is potentially useful for cancer staging and treatment. Evidence exists that DECT iodine mapping and virtual unenhanced images can replace conventional single-source CT cancer imaging. The improved visualization of iodine on low-energy images and the ability to “surf” to energy-level images that best enhance tumor-to-background contrast should improve tumor detection. For example, 50- to 52-keV DECT images reportedly yield the highest image contrast-to-noise ratio for visualization of hypodense masses.

**Figure 7. Acute gout in toes.**

of tumor vascularization and improve characterization of tumors’ relationships with adjacent vasculature and detection of tumors’ lymph node involvement because of low-energy images’ improved contrast between lymph tissue and vasculature. Less vascularized lesions can be differentiated from well-vascularized tumors. Repeated iodine-mapping DECT imaging might prove to be useful in monitoring tumors’ responses to treatment over time. Together, these facets of DECT imaging should allow sensitive detection, characterization, and mapping of the full extent of a tumor’s spread and individual tumors’ functional vascularization over time.

Lung nodules can be differentiated from malignant lung cancer with high sensitivity, and the effects of lung cancer on pulmonary function can be assessed using perfusion and ventilation imaging techniques. DECT PE imaging also is useful in monitoring lung cancer patients for this common treatment complication.

DECT plays an emerging role in liver and kidney cancer imaging, as well. Monoenergetic DECT images can improve tumor detection when liver tumors are well vascularized, particularly using low-energy data (e.g., 80 kV). Attenuation or spectral curves can be used to differentiate enhancing kidney tumors from hyperattenuating renal cysts.

Limitations of DECT

DECT is clearly useful and promising, but it is a relatively young imaging modality and more research is needed on many fronts to clarify its clinical applications. Despite important improvements regarding the early sequential-scan approaches to DECT, low-energy data sets sometimes suffer from poor contrast-to-noise ratios and yield higher image noise on virtual unenhanced images of genitourinary organs than is seen on unenhanced images in conventional contrast-enhanced CT scan series. Also, obese patients might require higher radiation doses to achieve sufficient image quality.

Virtual unenhanced DECT images can delete small kidney stones in iodine-subtraction postprocessing, and incomplete iodine subtraction can cause false-positive findings on virtual unenhanced images.

DECT data sets are significantly larger than those typically seen in traditional single-energy CT scans, as Schoepf and Colletti pointed out:

Indeed, the infinite variety of dual energy weighted average, iodine distribution and effective atomic number displays creates contrast possibilities akin to those of MRI, with associated challenging data management quandaries.

However, computing capacity, speed, and storage are improving rapidly and dramatically and are unlikely to represent meaningful or lasting barriers to adoption of high data-demanding DECT scan acquisition and postprocessing.

Outlook and Future Prospects

The evolution of DECT technology is difficult to predict, but significant technological innovations are being developed on several fronts. These include x-ray sources with more narrowly defined spectra and improved contrast-to-noise ratios, new detector designs, and contrast agent combination examinations employing iodine and gadolinium, tungstate, or tantalum for simultaneous oral and intravenous contrast enterography.

In theory, scans also can be performed at more than 2 energy levels, and multispectral CT imaging might come into clinical use in the future. However, it is far from clear that the clinical benefits of doing so with contemporary detector technology would justify the increased radiation doses. Multienergy CT using 4 or more simultaneous energy spectra also would cause technical challenges related to spectral overlap. However, quantum-counting detector technology is under investigation that could in the future allow conversion of each arriving photon into a quantifiable current, literally allowing photon counts and sorting by photon energy levels. Unfortunately, although promising for small lab animal scanning, prototype quantum-counting detectors are rapidly overwhelmed by incoming photons and are not yet viable for clinical applications with humans.

Another important front in DECT imaging is reducing metal implant–associated artifacts. As the elderly population grows, hip prostheses and other large metal medical implants will become more common, posing a challenge for traditional diagnostic imaging modalities. DECT has shown considerable promise in reducing the magnitude and interpretive effects of metal-associated imaging
artifacts. However, because the composition of hip prostheses and other medical implants varies and implant position relative to targeted anatomies also affects radiographic artifacts, more research is needed to identify and validate optimal scan parameters, extrapolated energies achieved via weighted blending of high- and low-energy DECT acquisition data sets, and other techniques for imaging tissues near implants.

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1. Dual-energy computed tomography (DECT) yields enhanced image contrast resolution compared with traditional single-energy CT imaging.
   a. true
   b. false

2. DECT imaging now is used routinely for which applications?
   1. perfusion imaging
   2. kidney stone characterization
   3. uric acid crystal visualization
   a. 1 and 2
   b. 1 and 3
   c. 2 and 3
   d. 1, 2, and 3

3. DECT was first described as a promising imaging technique in the:
   a. 1960s.
   b. 1970s.
   c. 1980s.
   d. 1990s.

4. Which factors hampered the first efforts to bring DECT into routine clinical use?
   1. high levels of image noise
   2. delays between consecutive scans
   3. anatomical registration errors
   a. 1 and 2
   b. 1 and 3
   c. 2 and 3
   d. 1, 2, and 3

5. Problems associated with single-energy helical CT scanners include:
   1. contrast enhancement timing.
   2. cardiac motion artifacts.
   3. respiratory motion artifacts
   a. 1 and 2
   b. 1 and 3
   c. 2 and 3
   d. 1, 2, and 3
6. The typical 2-tube DECT low-energy setting appropriate for improving image quality for obese patients is ______ kV.
   a. 40  
   b. 60  
   c. 80  
   d. 100  

7. Which innovation removes low-energy quanta from the higher spectrum to improve DECT image contrast without increasing radiation dose to patients?
   a. layered detector designs  
   b. improved field of view  
   c. tin filters  
   d. postprocessing software  

8. Which of the following systems has a slower gantry rotation time?
   a. DSCT  
   b. rapid kilovoltage-switching  
   c. dual-layer  
   d. single-source  

9. X-ray attenuation depends on which phenomena of interaction?
   1. photoelectric effect  
   2. Compton scatter  
   3. Rayleigh scatter  
   a. 1 and 2  
   b. 1 and 3  
   c. 2 and 3  
   d. 1, 2, and 3  

10. Photoelectric effect reflects the interactions of photons with atoms’:
    a. inner electrons.  
    b. outer electrons.  
    c. neutrons.  
    d. nuclei.  

11. Only materials composed of elements with relatively large ______ value differences can be differentiated using DECT technology.
    a. K-edge  
    b. Z  
    c. n  
    d. scatter  

12. Elements with low ______ values are most affected by Compton scatter.
    a. K-edge  
    b. Z  
    c. n  
    d. scatter  

13. The highest tube voltage typically used for DECT scans is usually no higher than ______ kV.
    a. 80  
    b. 100  
    c. 120  
    d. 140  

14. According to the article, comparing radiation doses across studies is complicated by differences in:
    1. protocols.  
    2. scan parameters.  
    3. DECT scanner equipment.  
    a. 1 and 2  
    b. 1 and 3  
    c. 2 and 3  
    d. 1, 2, and 3  

15. Compared with conventional single-tube CT scans, dual-source CT scans are associated with twice the radiation dose, on average, for a given imaging examination.
    a. true  
    b. false
16. Which type of scanner have most radiation dose studies employed?
   a. DSCT
   b. rapid kilovoltage-switching systems
   c. dual-layer systems
   d. gemstone

17. Which of the following algorithmic steps are performed to build clinically useful images?
   1. registration
   2. segmentation
   3. differentiation
   a. 1 and 2
   b. 1 and 3
   c. 2 and 3
   d. 1, 2, and 3

18. In DECT imaging, segmentation is largely a matter of:
   a. surface identification.
   b. constituent quantification.
   c. material decomposition.
   d. anatomic differentiation.

19. DECT weighted blending algorithms typically weight ______ % of a resulting image from the high-energy data set.
   a. 30
   b. 50
   c. 70
   d. 90

20. A single contrast-enhanced DECT acquisition generates scan data equivalent to ______ conventional CT perfusion scan(s).
   a. 0.5
   b. 1.0
   c. 1.5
   d. 2.0

21. Quantification algorithms visualize the physical densities of substances such as arterial plaques’ ______ components.
   a. iron
   b. calcium
   c. iodine
   d. fat

22. Iodine mapping can visualize the distribution of lung tissue:
   a. ventilation.
   b. hypertension.
   c. perfusion.
   d. saturation.

23. Contrast media bolus-triggered perfusion scan acquisitions are available for most contemporary DECT scanners.
   a. true
   b. false

24. How many seconds should the postinjection scan delay be for DECT pulmonary perfusion imaging?
   a. 3
   b. 5
   c. 7
   d. 9

25. Which of the following should be monitored for several minutes after a DECT lung ventilation scan is completed?
   1. respiration rate
   2. blood oxygen saturation
   3. blood pressure
   a. 1 and 2
   b. 1 and 3
   c. 2 and 3
   d. 1, 2, and 3
26. DECT’s ability to differentiate fatty liver from other tissues appears to be compromised in the presence of:
   a. alcohol.
   b. hemochromatosis.
   c. iodine.
   d. cirrhosis.

27. Early findings suggest DECT might improve participation in colorectal cancer screening by eliminating the need for:
   a. repeat examinations.
   b. biopsies.
   c. fecal tagging.
   d. bowel distention.

28. One problem with conventional CT imaging around metal implants is:
   a. beam hardening.
   b. photon enrichment.
   c. ring artifact.
   d. misregistration.

29. Which postprocessing method can reduce artifact effects associated with metal implants?
   a. curved planar maximum-intensity projection
   b. weighted blending of high- and low-energy data sets
   c. virtual enhancement
   d. surface rendering

30. The use of iodine contrast with DECT for oncology imaging improves which of the following?
   1. visualization of tumor margins
   2. characterization of tumors’ relationships with adjacent vasculature
   3. detection of tumors’ lymph node involvement
   a. 1 and 2
   b. 1 and 3
   c. 2 and 3
   d. 1, 2, and 3