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Evolution in Computed Tomography: The Battle for Speed and Dose.

Michael Lell, Joachim E. Wildberger, Hatem Alkadhi, John Damilakis ...+1 more authors

Institutions: Maastricht University, University of Zurich, University of Crete, German Cancer Research Center

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Evolution in Computed Tomography
The Battle for Speed and Dose

Michael M. Lell, MD, Joachim E. Wildberger, MD,† Hatem Alkadhi, MD,‡ John Damilakis, PhD,§ and Marc Kachelriess, PhD||*

Abstract: The advent of computed tomography (CT) has revolutionized radiology. Starting as head-only scanners, modern CT systems are now capable of performing whole-body examinations within a couple of seconds in isotropic resolution. Technical advancements of scanner hardware and image reconstruction techniques are reviewed and discussed in their clinical context. These improvements have led to a steady increase of CT examinations in all age groups for a number of reasons. On the one hand, it is very easy today to obtain whole-body data for oncologic staging and follow-up or for trauma imaging. On the other hand, new examinations such as cardiac imaging, virtual colonoscopy, gout imaging, and whole-organ perfusion imaging have widened the application profile of CT. The increasing awareness of risks associated with radiation exposure triggered the development of a variety of dose reduction techniques. Effective dose values below 1 mSv, less than the annual natural background radiation (3.1 mSv/year on average in the United States), are now routinely possible for a number of dedicated examinations, even for coronary CT angiography.

Key Words: computed tomography, dose reduction, iterative reconstruction, dual-energy CT, metal artifact reduction, perfusion CT, dynamic contrast-enhanced CT, detector technology, CT angiography, low kV

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Sir Godfrey Hounsfield, a British electrical engineer at EMI, and Alan MacLeod Cormack, a South African–born physicist were jointly awarded the Nobel Prize in Medicine in 1979 for their contributions to developing computed tomography, although they had worked independently from each other.[1] Johann Radon, an Austrian mathematician, had published much of the theoretical foundation of computed tomographic (CT) image reconstruction as early as 1917 (Radon transform).[2] In 1971, the first patient was imaged at Atkinson Morley's Hospital, London. The new insights into brain morphology possible with CT in patients received immediate worldwide scientific and media attention. This point in time is considered the beginning of a new era in clinical neuroscience. Early (first-generation) CT systems used a translate/rotate method, where the x-ray source, that only generated a pencil beam, and the detector moved laterally to cover the complete field of view, followed by a 1-degree rotation of the gantry. This process continued until 180 degrees were scanned. Such systems required 4 to 5 minutes to obtain the data for a single axial slice. Second-generation CT systems using a small x-ray fan beam were introduced in 1972; they still used the translate/rotate method. Both first- and second-generation systems were head-only scanners. Third-generation systems with a full x-ray fan beam that covered the complete field of view, and a simultaneously rotating tube and detector array were introduced in 1975. The geometry of modern CT systems is still similar to this setup. Fourth-generation systems with a full 360-degree detector ring and a rotating x-ray tube as well as electron beam CT were eventually both replaced by further developments of third-generation-type CT systems. In 1987, continuously rotating systems based on slip-ring technology were introduced with a gantry rotation time of 1 second. Spiral or helical CT was introduced in 1989[3] and led to a breakthrough in truly 3-dimensional (3D) CT imaging and in CT angiography (CTA). Cardiac CT was already announced in the 1970s[4,5] but was based on dedicated systems such as electron beam CT[6] and the dynamic spatial reconstructor.[7] It finally entered clinical routine and became widely available by combining the spiral trajectory on standard third-generation systems with a dedicated reconstruction approach.[8,9] In addition to the development of larger detector arrays with more detector rows, dual-source CT (DSCT) was introduced in 2005 with the aim of further increasing scan speed and double temporal resolution. High scan speed and temporal resolution had a strong impact on the technical development of CT and was particularly important for cardiac CT. The topic of cardiac CT is more completely covered in Wintersperger et al.[10]

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From the *Department of Radiology, University Hospital Erlangen, Erlangen, Germany; †Department of Radiology and Nuclear Medicine, Maastricht University Medical Center, Maastricht, the Netherlands; ‡Institute for Diagnostic and Interventional Radiology, University Hospital Zurich, Zurich, Switzerland; §Department of Medical Physics, University of Crete, Iraklion, Crete, Greece; and ||X-Ray Imaging and CT, German Cancer Research Center (DKFZ), Heidelberg, Germany.
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Correspondence to: Michael M. Lell, MD, Department of Radiology, University of Erlangen-Nuremberg, Ulmenweg 18, 91054 Erlangen, Germany. E-mail: michael.lell@uk-erlangen.de.

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SLICE WAR

In medical imaging, the *x-y* plane denominates the patient's cross section, whereas the *z*-direction denominates the longitudinal axis. A typical detector array consists of 800 to 1000 detector elements in-plane and of up to 320 detector rows in the *z*-direction. The general advantage of more rows is the simultaneous acquisition of more data along the patient's longitudinal axis, which mainly allows increased scan speed. In 1992, Elscint introduced the first third-generation CT system with two detector rows. With the availability of solid-state detectors based on scintillating ceramics,[11] the other CT vendors followed with 4-row detector arrays in 1998. Since then, a steady race for the highest number of active detector rows (or later slices per rotation) broke out, culminating in 2007 in 320 detector rows in the Toshiba Aquilion One (Table 1). The number of detector rows and slices acquired per rotation were equivalent until the introduction of 64-slice scanners. Whereas General Electric (GE), Philips, and Toshiba mounted 64-detector row panels, Siemens went a different way. Siemens implemented a double read-out technique called double *z*-sampling, which uses a periodic motion of the focal spot not only in the *x-y* (to double the number of rays per detector row) but also in the *z*-direction with the goal of doubling the number of independent slices acquired in the *z*-direction.[12,13] Philips later shared this concept when introducing their 256-slice system. Although the number of slices nominally doubles with this technique, the total width of the detector panel (and number of detector rows) does not change. Challenges associated with large detector panels are increased scatter, cone beam artifacts, heel effect, and a potential tradeoff in image quality.[14] On the other hand, broader detector panels cover more anatomy in a single rotation, optimally the whole organ of interest.

TABLE 1. Data Sheet of Up-to-Date High-End CT Systems

CT System	Vendor	Configuration	Collimation (mm)	Cone (Degree)	Rotation (Second)	Max Power
Revolution CT	GE	256 × 0.625 mm Gemstone Clarity	160	15	0.28	103 kW Performix HDw
Brilliance ICT	Philips	2 · 128 × 0.625 mm NanoPanel^{3D}	80	7.7	0.27	120 kW iMRC
IQon	Philips	2 · 64 × 0.625 mm NanoPanel Prism	40	3.9	0.27	120 kW iMRC
Definition Edge	Siemens	2 · 64 × 0.6 mm Stellar	38.4	3.7	0.28	100 kW Straton
Definition Flash	Siemens	2 · 2 · 64 × 0.6 mm Stellar	38.4	3.7	0.28	2·100 kW Straton
Somatom Force	Siemens	2 · 2 · 96 × 0.6 mm Stellar^{Infinity}	57.6	5.5	0.25	2·120 kW Vectron
Aquilion ONE Vision	Toshiba	320 × 0.5 mm Quantum	160	15	0.275	100 kW MegaCool Vi

Names of detector and x-ray tube systems in Table 1 are trademarks. The notation of the detector configuration is the number of active detector rows × the slice thickness. An additional factor of 2 indicates that a *z* flying focal spot is used to double the number of slices. Another factor of 2 indicates that a dual-source dual-detector configuration is implemented. The collimation refers to the active width of the detector. All length values are scaled to the isocenter (*z*-axis).

Such detector panels facilitate interventional CT-guided procedures and expands the coverage in dynamic ("perfusion") CT.

GANTRY TECHNOLOGY

The introduction of slip-ring technology for the transport of data and energy to and from the gantry enabled continuous rotation of x-ray tube and detector. This was an essential prerequisite for the development of spiral CT scanners.[3,15] Most recent CT systems with gantry rotation times down to 0.25 s use noncontact data and energy transfer between the rotating and stationary parts of the gantry and some of them even friction-free air bearings. Tremendous centrifugal forces in the order of 40 g act on the rotating hardware, which weighs up to 1 ton and more. This has a significant impact on the x-ray tube design for mechanical reasons. Larger patient body sizes and the desire to perform whole-body scans also demand new patient table technology. To realize fast and particularly high-pitch scanning (table speeds of up to 74 cm/s are currently in use) the acceleration and deceleration range of the table needs to be taken into account to assure patient's comfort. The capacity of the CT table has been adapted to increasing patient's body weight, with special tables allowing 300 kg and more. Larger patients require not only increased table capacity but also greater gantry diameter (up to 90 cm in dedicated systems), extended reconstruction field of view, and higher tube output. In addition, a larger bore diameter provides more space for CT-guided interventions.

TUBE TECHNOLOGY

Despite the permanent struggle to reduce radiation exposure, increasingly powerful x-ray tubes have been developed to meet clinical needs. To perform a CT scan of chest and abdomen or an aortic CTA in heavy patients within a scan time short enough for the patients to comfortably hold their breath requires faster rotation times and table speeds, and therefore higher tube output is necessary. Typical maximum tube power is in the order of 100 kW or more. With dual-source systems, these values double. Since scan times significantly decreased during the past decades, the latest generation of x-ray tubes is optimized for very short exposure times at high exposure levels (Fig. 1A). Among vendors, a significant variation in maximum tube power performance can be observed (Fig. 1B).

Higher tube power does not mean higher patient's dose, but in contrast, it is potentially the basis of measures that significantly reduce the patient's dose. More x-ray power for instance allows using stronger prefiltration, which removes undesired low-energy photons from the spectrum, photons that would otherwise mainly contribute to the patient's dose but not to the CT image.

Higher output is also required if low-kilovolt scanning (see the "dose reduction techniques-low-kilovolt scanning" section below) is performed. Whereas the tube current can be selected freely in a relatively wide range, the tube voltage is typically limited to a few settings, although some recent systems allow operation at voltages ranging from 70 kV to 150 kV at 10 kV increments.[16–26] A flexible choice of tube voltages, including the possibility to go below 80 kV, is of importance for patient's dose reduction and contrast optimization and, in particular, for pediatric examinations.

In addition to allowing higher tube power values, x-ray tubes have also been optimized with the aim of minimizing cooling delays. This is possible by introducing direct (or active) cooling instead of using indirect cooling. One example of direct cooling is rotating envelope tubes where the anode is in contact with the envelope and where the envelope is in contact with a cooling medium: anode, cathode, and housing rotate together as a unit.[27] Another implementation of active cooling is to use spiral groove bearing technology, which replaces the conventional ball bearings and where liquid metal is used not only for lubrication (nonabrasive contact) but also to actively cool the anode.[28]

DETECTOR TECHNOLOGY

Since the late 1990s, CT detectors are indirect converters, where the x-ray energy is first converted into visible light, which then is captured by a photodiode and converted into an electric current (Fig. 2, left). Indirectly converting detectors in clinical CT systems are based on gadolinium oxysulfide (Gd_2O_2S, also called GOS or Gadox) ceramic scintillators. The scintillator material is optimized for high-dose usage, for high light output, for low crosstalk, and for low afterglow or ghosting. During the past years, detectors were further optimized with respect to electronic noise. This noise is intrinsic to the detector, showing up especially in the dark image (no x-ray exposure). Electronic noise is a noise source in addition to the x-ray quantum noise. Whereas quantum noise cannot be avoided, electronic noise can be reduced by decreasing the distance between the photodiode and the analog-to-digital converter electronics. This is achieved by designing more compact electronics that can be placed closer to the detector pixels and that require shorter analog wires. The latest step includes fully integrated electronics that have become possible with new contacting technology, the so-called through silicon vias.[29,30] With such a technology, the analog distance can be reduced to a few millimeters. Less electronic noise implies significantly lower image noise in those situations, where only few x-ray quanta reach the detector (obese patients, or very low milliampere second [mAs] settings as desirable for pediatric scans or screening programs).

Currently, semiconductor-based direct converting detectors are under development for CT imaging (Fig. 2, right).[31–33] With such a technology, a detected x-ray photon generates a very short pulse, short enough to count each photon. In addition, the area under the pulse, and thus the pulse height, is proportional to the photon energy. The aim is to design these photon-counting detectors with energy discrimination capabilities based on pulse height analysis to gather spectral x-ray information without the need to apply to different x-ray tube

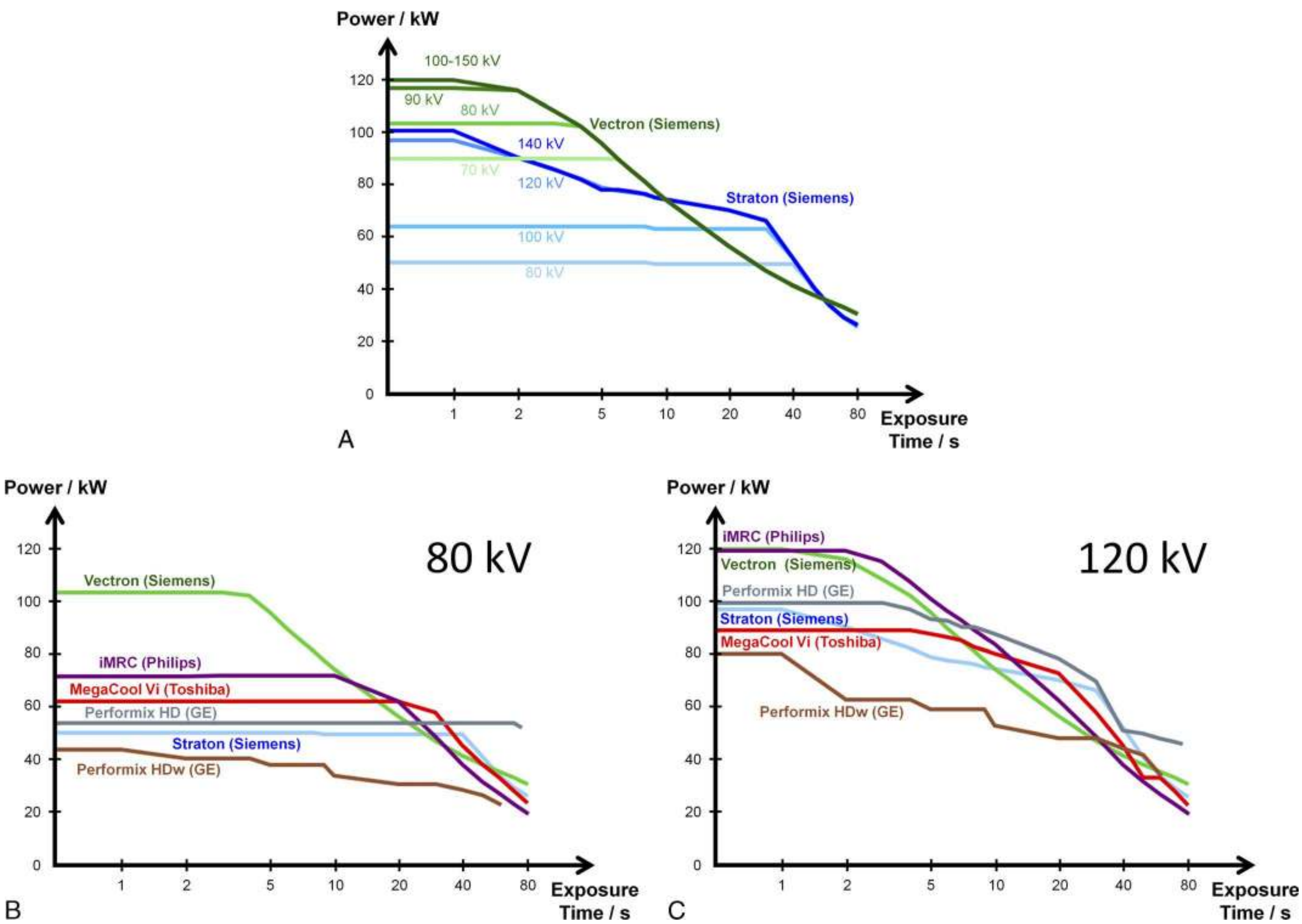

FIGURE 1. The available tube power is a function of the desired exposure time. The maximal tube power (eg, 120 kW) is available only for very short scans to avoid thermal damage to the anode. A, Two of the tubes are from the same vendor (blue, second last generation; green, latest generation). The trend to focus on shorter exposure times with high power levels is clearly visible. B, Tube performance across vendors at 80 kV. C, Tube performance across vendors at 120 kV. Plotted values are those available for routine spiral scanning at the scanner's console. Figure 1 can be viewed online in color at www.investigativeradiology.com.

voltages. These detectors are expected to improve tissue contrast and reduce image noise.[33] The latter is possible not only because the photon counters have zero electronic noise but also because the energy bin measurements can be combined in a statistically optimal way.[34] Dose reduction,[35] decreased beam hardening artifacts,[34] accurate K-edge imaging,[36,37] simultaneous multicontrast agent imaging,[38,39] and true quantitative imaging[40] are further potential benefits of this new detector generation. However, as of today, the technology is not yet mature, and only a few experimental systems exist that are not (yet) approved for patient examinations.

DOSE REDUCTION TECHNIQUES

Tube Current Modulation/Automatic Exposure Control

The technical basis for tube current adaption dates back to 1981.[41] Angular tube current modulation (TCM) resulted in 15% to 50% dose reduction, depending on the anatomical region in the x-y plane.[42] Online tube current modulation did not only reduce patient exposure but also homogenized noise distribution and therefore improved image quality.[43] The logical advancement of in-plane TCM was longitudinal or *z*-axis TCM. In analogy to angular modulation that considers different attenuation, in-plane (transverse projection vs anteroposterior projection at the level of the shoulders) longitudinal TCM aims to homogenize noise, taking into account the different attenuations, for example, of the chest as compared to the abdomen or pelvis. Different solutions from the major vendors became available using either a sinusoidal or attenuation-based online modulation algorithm. Automatic exposure control (AEC) consists of a group of algorithms, which incorporate (3D) TCM and aim to deliver a predefined image quality across a range of patient sizes derived from the topogram, increasing the volume CT dose index ($CTDI_{vol}$) for large and decreasing $CTDI_{vol}$ for small patients. Because AEC algorithms assume that the patient center is in the isocenter, correct centering is of importance. If the scan range goes beyond the range of the topogram, CT systems may act differently using either maximum or minimum mAs setting or something in between (standard mAs setting or mAs setting at the last calculated position). The image quality is defined differently for CT systems of different manufacturers: GE's AutomA uses a noise index that refers to the standard deviation of the CT value within a specific water phantom, which is converted to the individual patient; Philips' DoseRight uses a reference image, whereas Siemens' CareDose 4D uses reference mAs that defines the image quality in a "standard adult" of 70 kg body weight. Toshiba's $^{\mathrm{SURE}}$Exposure 3D offers at least three quality settings based on the target standard deviation of image noise. To allow comparison of protocols between different platforms,

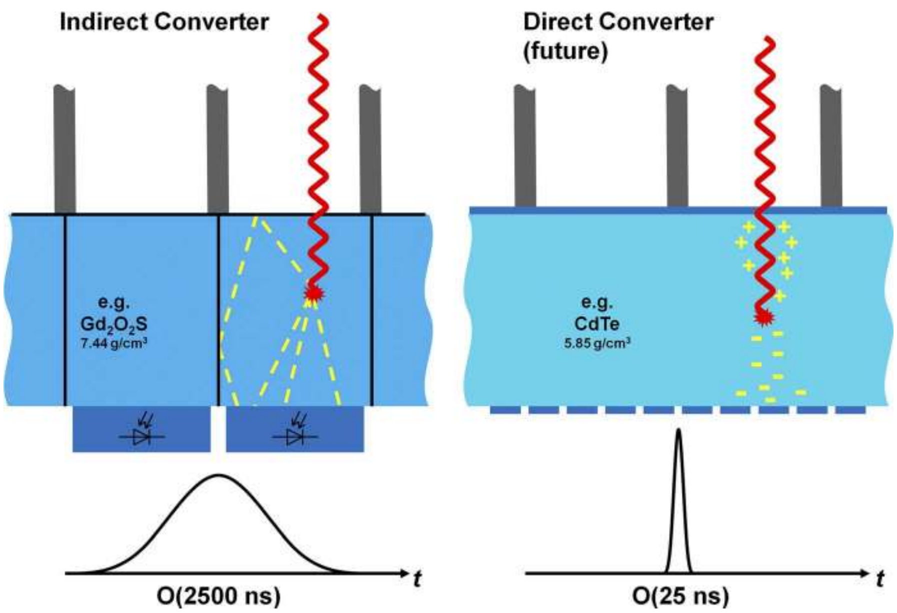

FIGURE 2. Conventional CT detectors (left) and future photon counters (right). Photon counting becomes possible owing to the very short signal peak generated by each x-ray photon, such that adjacently arriving photons typically generate two separate peaks. This is not the case for the indirect converters. Figure 2 can be viewed online in color at www.investigativeradiology.com.

not only the AEC setting or noise level but also the milliampere second and $CTDI_{vol}$ should be given.

Electrocardiogram (ECG)-correlated TCM reduces the tube output in phases of the cardiac cycle not intended for image reconstruction in retrospectively ECG-gated cardiac CTA. The shorter the time window with full tube output, the higher the dose reduction for the patient, at the cost of limited flexibility in image reconstruction of high-quality CTA data. Image reconstruction in those phases with very low x-ray exposure is still possible, which maintains the option to perform functional ventricular assessment.[44–48] A similar technique is available for prospectively gated scans, where the gating window can be chosen larger than necessary, and the full tube current will only be given to the minimum required range.

Dynamic Collimators

Detector panel width and pitch influence the *z*-overscanning effect, which is inherent to spiral CT. Spiral image reconstruction requires data from above and below each image position; therefore, at least one additional half-rotation (180 degrees in parallel ray geometry) is necessary at both ends of the spiral scan. As a result, additional tissue is exposed to radiation outside the imaged volume. For single–detector row scanners, *z*-overscanning may be considered irrelevant. However, the effect increases with the number of detector rows and becomes significant with large area detectors and cone-beam geometry. To minimize the *z*-overscanning effect, special dynamic collimators have been introduced that asymmetrically open and close at the edges of the scan range.[49]

Low-Kilovolt Scanning

Scanning at low-kilovolt setting increases the attenuation of contrast material (CM). For example, iodine attenuation increases by a factor of 1.97 (70 kV) and 1.44 (90 kV), respectively, compared with the one at 120 kV. Thus, the iodine dosage can be reduced by roughly 50% (70 kV) and 30% (90 kV) while maintaining identical attenuation.[19] The increased iodine attenuation can be used to reduce either the volume of CM,[50–52] or the radiation exposure (compensating higher image noise by higher contrast), or a combination of both. Operating the CT system at low-kilovolt at a fixed tube current minimizes the x-ray exposure of the patient but also increases image noise because fewer photons reach the detectors. In clinical practice, a change of the tube voltage setting requires a simultaneous adjustment of the tube current to keep image quality high. Therefore, low-kilovolt scanning, historically, was not popular in CT, except for perfusion CT, although the benefits in cardiac CTA,[21,53–57] aortic CTA,[58–60] and pediatric CT[23,61–63] have been demonstrated. Automatic selection of the tube voltage and adaption of the tube current, using information on the patient's attenuation from the localizer scan, and accounting for the planned examination type (nonenhanced scan, contrast-enhanced parenchymal scan, and CTA) translated this technology into routine use. Depending on the examination type, a dose reduction between 10% and 30% is possible.[16,17,25] The restricted x-ray tube output limited the automated selection of 80 kV or 100 kV in larger patients or with very fast scan modes. The recent availability of new high-performance x-ray tubes that provide very high tube currents at low tube voltages, however, makes low-kilovolt scanning routinely possible for a wide range of patients. Additional prefilters are in use for some protocols to remove undesired low-energy radiation from the low-kilovolt spectrum to maximize image quality and to minimize patient dose down to the level of conventional radiography.[18,22,26]

FAST SCANNING

The scan speed increases with an increased number of detector rows, although in some implementations, not all detector rows are available for spiral scanning. For dual-source CT systems, there is a special high-pitch mode, which is a very fast spiral scan mode that uses pitch values up to 3.2 to 3.4 and table speed of up to 737 mm/s. The data gaps occurring with single-source CT systems at pitch values greater than 1.5 are closed by the data from the second source in DSCT systems. Since the cone angle of DSCT is only half as large as the cone angle of single-source systems of twice the detector width, the same scan speed can be achieved with less cone beam artifacts (with the remaining cone beam artifacts being corrected by the iterative image reconstruction). The high pitch mode was originally designed to improve cardiac CTA.[53,56,64–76] All CT data of the heart are acquired in a fraction of a single heartbeat (usually late diastole); thus, compared to a conventional low-pitch scan mode for cardiac CT with retrospective gating, no redundant data are acquired, which makes the scan mode very dose effective. In CT examinations that use all data for image reconstruction, the dose reduction effect vanishes, equal dose distribution within the scan range assumed.[61,66] Powerful tubes (Table 1) are necessary to deliver the required dose within that very short time, especially if low-kilovolt scanning is performed. Axial scanning with a large detector panel (up to 160 mm) covering the object of interest (eg, heart and brain) is an alternative[77–81] that does not require fast table speed with acceleration and deceleration but is limited in the length of *z*-coverage and faces some severe image quality challenges.[14]

Fast scanning offers a variety of advantages in daily practice; for example, the option to perform dose-effective ECG-triggered chest CTA ("triple-rule-out"),[58,82–85] to reduce the amount of CM in patients with renal impairment,[50,86–88] and to achieve a very homogeneous enhancement pattern. Scan times in the order of a second or less effectively help to suppress motion, for example, in patients who are not able to hold their breath or are unable to cooperate (small children, unconscious patients, severely impaired patients). In many cases, sedation is not necessary any more (Fig. 3).[20,58,61,89,90]

IMAGE RECONSTRUCTION

Filtered back projection (FBP) has been the reconstruction algorithm of choice during the recent decades. Filtered back projection is an analytical image reconstruction algorithm, that is, an inversion formula for a simplified measurement model. Filtered back projection is a reliable and robust algorithm that can be implemented with high computational performance. It has desirable properties such as linearity and translational invariance that make image quality easily understandable and facilitate image quality assessment (the image of the sum of two objects equals the sum of the object's individual images). However, owing to the simplicity of the measurement model, FBP cannot optimally account for a number of physical effects, such as the photon statistics, beam polychromaticity, or finite width x-ray beams. To account for the photon statistics empirical sinogram restoration approaches, so-called adaptive filters have been used for decades. With the introduction of multirow detector CT, the efficiency of such adaptive filtering improved tremendously owing to the possibility of filtering across detector rows or even across projections.[91] Adaptive filtering approaches are still in use today, and not just in combination with the FBP algorithm.

During the past decade, new classes of image reconstruction algorithms have become commercially available. Iterative image reconstruction approaches have been in use for a long time in nuclear medicine, but the lack of sufficient computing power prohibited their early implementation in CT (although Hounsfield's CT system was actually running an iterative reconstruction algorithm). Iterative reconstruction allows making use of more complicated measurement models than FBP and thereby promises images of lower artifact content and lower image noise. In contrast to FBP, this inversion is not based on finding a reconstruction formula but rather on iteratively estimating an image that best fits to the acquired raw data. This iterative estimation requires a series of iterations, however. Every iteration step consists of a forward projection, followed by a comparison of the forward-projected data with the measured raw data, followed by a back projection. This explains its high computational demands. Although a long list of publications focusing on iterative reconstruction techniques exists, insufficient data are published about the specific algorithmic implementation of the commercially available solutions.

The first iterative image reconstruction algorithms that became available commercially-adaptive statistical iterative reconstruction (ASIR; GE), adaptive iterative dose reduction (AIDR) and AIDR 3D (Toshiba), iterative reconstruction in image space (IRIS; Siemens), and iDose (Philips)-were purely image based (Fig. 3). These image postprocessing algorithms can be thought of being edge-preserving filtering, aiming at noise reduction in homogeneous regions and at preservation or improvement of spatial resolution at edges. In fact, the term *image restoration* would be more appropriate for these types of algorithms than the term *image reconstruction*.

In the meantime, vendors have provided "fully iterative image reconstruction algorithms" in the sense that the reconstructed image undergoes a forward projection to be compared with the measured raw data with the aim of minimizing the residual errors. This procedure is necessary to minimize artifacts such as cone beam or spiral artifacts. These raw data iterations can also be used to minimize image noise (for dose reduction) and to model the finite width and the exact shape of the x-rays (for resolution recovery). However, it can be shown that if an exact inverse of the forward projector exists, the iterative process can be converted into one that entirely operates in the image domain.[92] Furthermore, it can be shown that the effect of ray modeling in CT can also be converted to an image domain operation, given that this filtering is applied to a master image with rather small voxels.[93,94] These two observations justify minimizing the number of (computationally demanding) iterations between image domain and raw data domain in favor of performing more (computationally efficient) iterations in image domain (Fig. 4; Table 2).

Adaptive statistical iterative reconstruction (ASIR) is an iterative reconstruction algorithm of GE. Image noise is reduced by raw data preprocessing, FBP of the preprocessed raw data, modeling of the statistical system properties, and loop wise image regularization. The model incorporates statistical information from the CT system (including photon statistics and electronic noise) and details of the system optics (including the size of each detector cell, dimensions of the focal spot, shape, and size of each image voxel). The iterative procedure mainly involves the image data domain. The reconstruction time for this technique is only slightly longer than that for the FBP, therefore being well suited for the workflow of a busy imaging center. The

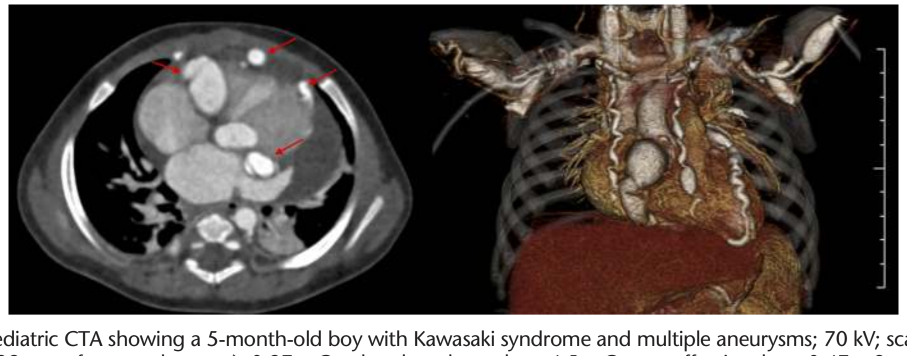

FIGURE 3. High-pitch pediatric CTA showing a 5-month-old boy with Kawasaki syndrome and multiple aneurysms; 70 kV; scan speed, 73 cm/s; pitch, 3.2; ADMIRE 2, $CTDI_{vol}$ (32 cm reference phantom), 0.27 mGy; dose length product, 4.5 mGy cm; effective dose, 0.47 mSv; no sedation required. Figure 3 can be viewed online in color at www.investigativeradiology.com.

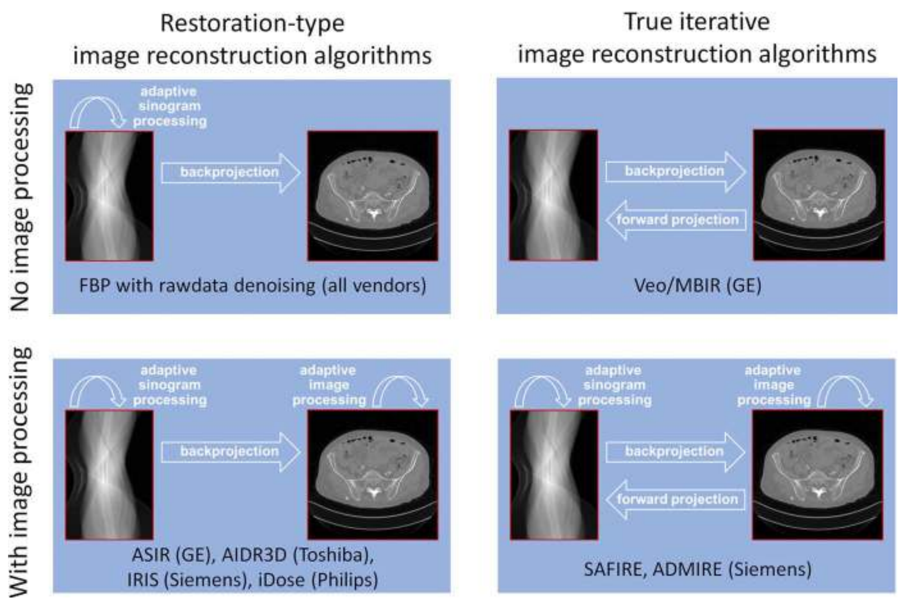

FIGURE 4. Classification of image reconstruction algorithms provided by the vendors (adapted from Kachelriess[95]). The algorithms are mostly a black box to the scientific community and users of the CT systems, with few details published. Two algorithms cannot be classified because the corresponding vendors do not disclose sufficient information (IMR [Philips], ASIR-V [GE]). Figure 4 can be viewed online in color at www.investigativeradiology.com.

regularization steps in the image domain affect not only the noise level but also the image texture, sometimes called "plastic-like" appearance. To improve acceptability and to restore the more familiar texture, ASIR images are typically linearly blended with FBP images. Zero percent ASIR corresponds to an FBP image, whereas 100% ASIR provides the highest level of noise reduction at the cost of an altered image texture. Increasing the blending ratio significantly decreases measured noise levels and increases signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR), but this does not necessarily translate into a significant improvement in diagnostic confidence and specificity.[96] In abdominal CT, weighting factors of 25% to 60% are frequently used.[96] With ASIR 40%, a dose reduction in abdominal CT of 25% has been realized.[97] In another study, body mass index (BMI) adapted dose reduction of 23% to 66% were reported, but the mean $CTDI_{vol}$ of the reference protocol in this study was very high (21 mGy for BMI < 20 kg/m^2 to 27 mGy for BMI > 25 kg/m^2).[98] Similar dose reduction potential was reported in cardiac CTA (44% dose reduction),[99] chest CT (25% dose reduction),[100] and head CT (31% dose reduction).[101]

Model-based iterative reconstruction (MBIR or VEO™, GE) is a computationally highly demanding "true" iterative reconstruction scheme; MBIR incorporates not only modeling of photon and noise statistics but also modeling of system optics.[26] Originally, blending with FBP images was not performed, and no specific reconstruction kernels were provided.[102–104] Dose reduction values of 50% and more without sacrificing image quality in abdominal CT were reported using MBIR.[105–108] When image qualities of abdominal CT were compared, MBIR images acquired at half (50%) standard-dose had less noise and better quality than half-dose ASIR and half-dose FBP, and also less noise and similar image quality compared with standard-dose FBP.[105] In another study of the abdomen and pelvis, subjective image quality of low-dose MBIR scans (76% dose reduction) was found to be superior compared to standard-dose FBP and ASIR.[107] In small liver lesions (<10 mm), detection and conspicuity were significantly higher with MBIR than with ASIR.[109] In low-dose chest CT, intraindividually comparing image quality of a 120 kV/50 mAs and 120 kV/4 mAs protocol, MBIR was criticized for blotchy pixelated appearance and reduced image sharpness compared to FBP. However, the performance of MBIR was significantly superior to that of FBP for the detection of non-calcified pulmonary nodules at the same dose level.[110] Submillisievert images ($CTDI_{vol}$, 2 mGy) were found to be diagnostically acceptable for the evaluation of lung parenchyma (not the mediastinum) even with FBP, but MBIR images at that dose were rated suboptimal because of

TABLE 2. Properties of the Iterative Reconstruction Techniques Used in CT Today

Vendor	Reconstruction Algorithm	Additional Parameters	Recon Time rel. to FBP	Sinogram Iterations	Image Iterations	Full Iterations
All	FBP	—	1	√	—	—
GE	ASIR, ASIR-V	0%–100% (eg, ASIR, 30%)	2	√	√	—
	MBIR/VEO	—	30–50	—	—	√
Philips	iDose	Levels 1–7	2	√	√	—
	IMR	Soft, Routine, or SharpPlus	5	?	?	?
Siemens	IRIS	Strength 1–5	1–2	√	√	—
	SAFIRE	Strength 1–5	1–2	√	√	√
	ADMIRE	Strength 1–5	1–2	√	√	√
Toshiba	AIDR, AIDR 3D	Mild, standard, or strong	1	√	√	—

The reconstruction time varies with different reconstruction hardware.

a loss of conspicuity of normal pulmonary structures despite showing the lowest image noise.[111] No significant differences in sensitivity between ASIR images ($CTDI_{vol}$, 1.77 mGy) and MBIR images ($CTDI_{vol}$, 0.39 mGy) were reported for ground glass opacities and nodules.[112] Filtered back projection, ASIR, and MBIR resulted in different emphysema indices and wall thickness measurements, which needs to be taken into account, when images from different sites and CT units are compared in the follow-up.[113] In head CT, MBIR was reported to be superior to ASIR in image quality and artifact reduction at equivalent dose level but at the cost of a mean reconstruction time of 32 minutes.[114] Higher resolution and lower noise have been reported at the cost of significantly longer reconstruction times (ranging from 10 to 90 minutes, depending on the number of slices to be reconstructed). Because the long reconstruction times significantly interfere with clinical workflow, a hybrid with ASIR (ASIR-V) is currently available instead of MBIR.

iDose 4 (Philips) is a hybrid iterative reconstruction algorithm with two denoising components: an iterative maximum likelihood-type sinogram restoration based on Poisson noise distribution and a local structure model fitting on image data that iteratively decreases the noise. The level of noise reduction in the images can be selected in 7 levels; the higher levels indicate greater noise reduction, also at the cost of increasing image texture alteration (blotchy pixilated appearance) compared to FBP images. Image reconstruction is fast enough not to delay the clinical workflow (approximately 20 frames per second).[115] Fifty-five percent dose reduction was reported using iDose in retrospectively ECG-gated coronary CTA using 256-multidetector computed tomography (MDCT).[116]

Iterative model reconstruction (IMR; Philips) is a more advanced algorithm, but very little detail has been published until now. The visibility of peripheral lung vessels on low-dose images ($CTDI_{vol}$, 1.8 mGy) reconstructed with iDose (levels 2 and 4) and standard-dose images ($CTDI_{vol}$, 5.6 mGy) reconstructed with FBP was considered optimal, but the visibility of small peripheral blood vessels on low-dose images reconstructed with IMR in a prototypic setting was compromised in some cases, although the overall noise level was lowest.[115] Intravascular noise reduction up to 88% in cardiac CTA was reported, with image noise amplitude within the left main coronary artery reduced by a factor of 1.3 from FBP to iDose 4 and by a factor of 2.6 from iDose 4 to IMR[117]; no correlation with invasive coronary angiography was performed, so the impact on stenosis grading could not be assessed.

Iterative reconstruction in image space (IRIS; Siemens) is an approach combining adaptive sinogram postprocessing, analytical image reconstruction, and iterative image processing. Similar to the other restoration type image reconstruction techniques, an initial FBP image containing all frequencies is used for subsequent iterative processing loops to enhance sharpness and reduce noise taking into account the physical properties of the scanner system and the reconstruction parameters. Repetitive correction loops account for noise reduction while preserving edge information and low-contrast structures.[118,119] The effect of IRIS could be elegantly assessed with DSCT, which allows acquisition of full-dose and half-dose CT data simultaneously by using data for image reconstruction from both or only one tube detector array. No statistically significant differences in diagnostic confidence or artifacts between full-dose FBP and 50% dose IRIS images of the abdomen were reported, except an altered image noise pattern with IRIS.[118] An identical setup was used to test the effects in chest CT with similar results.[120] Iterative reconstruction in image space significantly improved image quality compared to FBP images at the same dose level ($CTDI_{vol}$, 45 mGy) in head CT and at least matched the objective and subjective image quality parameters of standard FBP images ($CTDI_{vol}$, 60 mGy).[119]

Sinogram-affirmed iterative reconstruction (SAFIRE; Siemens) and the advanced modeled iterative reconstruction (ADMIRE; Siemens) are fully iterative. A few iterations between image and raw data domain are conducted to reduce artifacts. Noise reduction is conducted in sinogram and image domain. This combines high computational performance with high image quality.[92]

A similar DSCT approach as in reference[118] was used to assess image quality, diagnostic performance and the optimal strength setting of SAFIRE in abdominal CT. Sinogram-affirmed iterative reconstruction strength 2 (strength values ranging from 1 to 5 are available) at 50% dose was considered noninferior to FBP at full dose, although sensitivity for small lesions (<1 cm) was reduced using 50% SAFIRE (sensitivity, 55%) instead of 100% FBP (sensitivity, 70%; P = 0.08).[121] Acceptable image quality in cardiac CTA at very low dose (<0.1 mSv) became possible in selected patient groups using SAFIRE.[122] Chest CTA using low-kilovolt scan protocols and SAFIRE provided diagnostic image quality at low dose (<1.5 mSv).[123] Phantom (COPDGene 2 test object) measurements demonstrated accurate quantitative chest CT images with acceptable image noise at very low dose levels ($CTDI_{vol}$, 0.15 mGy) using ADMIRE.[124] Lower image noise, higher diagnostic confidence, and higher sensitivity for nodule detection was reported for ADMIRE compared to SAFIRE and FBP in low-dose chest CT (down to $CTDI_{vol}$, 0.14 mGy).[18]

The AIDR and AIDR 3D (Toshiba) are sinogram domain and image domain-based iterative reconstruction algorithms without a forward projection from image domain to sinogram domain.

METAL ARTIFACT REDUCTION

Artifacts from strongly attenuating objects like metallic implants, dental fillings, or highly concentrated contrast material (CM) are a problem frequently encountered in CT. They lead to severe image artifacts due to photon starvation, to beam hardening and, in particular, to scatter. Sometimes, such artifacts can be reduced using a physics-based approach.[125] In most cases, however, such severe artifacts require a dedicated metal artifact reduction (MAR) software.

The first algorithms for MAR, as well as many of the algorithms proposed since then are pure sinogram inpainting methods.[126,127] No matter how sophisticated the inpainting algorithm is, the original artifacts are removed, whereas significant new artifacts are introduced. The resulting images are rarely of diagnostic quality.

A decisive step toward diagnostic MAR images was the introduction of a so-called prior image, that is obtained from the initial uncorrected images by soft thresholding with the thresholds being set to represent the tissue classes air, soft tissue and bone, for example, with 3 predefined CT values. The normalized metal artifact reduction (NMAR) algorithm uses this prior image for normalization and denormalization. Forward projecting this prior image yields the prior sinogram, which is used to normalize the original raw data (Fig. 5). The normalized raw data are very homogeneous outside the metal trace; and thus, the metal trace can be safely replaced by linear interpolation or any other kind of inpainting approach. The inpainted raw data are then denormalized by pixelwise multiplication with the prior sinogram. The corrected raw data are reconstructed by FBP, the metal image is inserted, and, in some cases, high frequencies of the initial uncorrected images are added to obtain the final corrected images.[44,45] An iterative version of the NMAR algorithm, called IMAR is commercially being implemented on Siemens systems.[128] Another approach similar to the NMAR algorithm is the orthopedic metal artifact reduction (O-MAR) algorithm used commercially by Philips.[129] These new classes of MAR algorithms have been successfully used to improve image quality by suppressing metal artifacts in patients with dental[130–132] and orthopedic hardware.[133,134]

Virtual monochromatic imaging using dual-energy CT (DECT) (see the "Dual-Energy CT" section below) has been proposed as an alternative approach for MAR, but it has been demonstrated that DECT provides suboptimal image quality in patients with bilateral prostheses or in those with dental hardware.[135,136] Nevertheless, in cases where no dedicated MAR software is available, artifact reduction using such

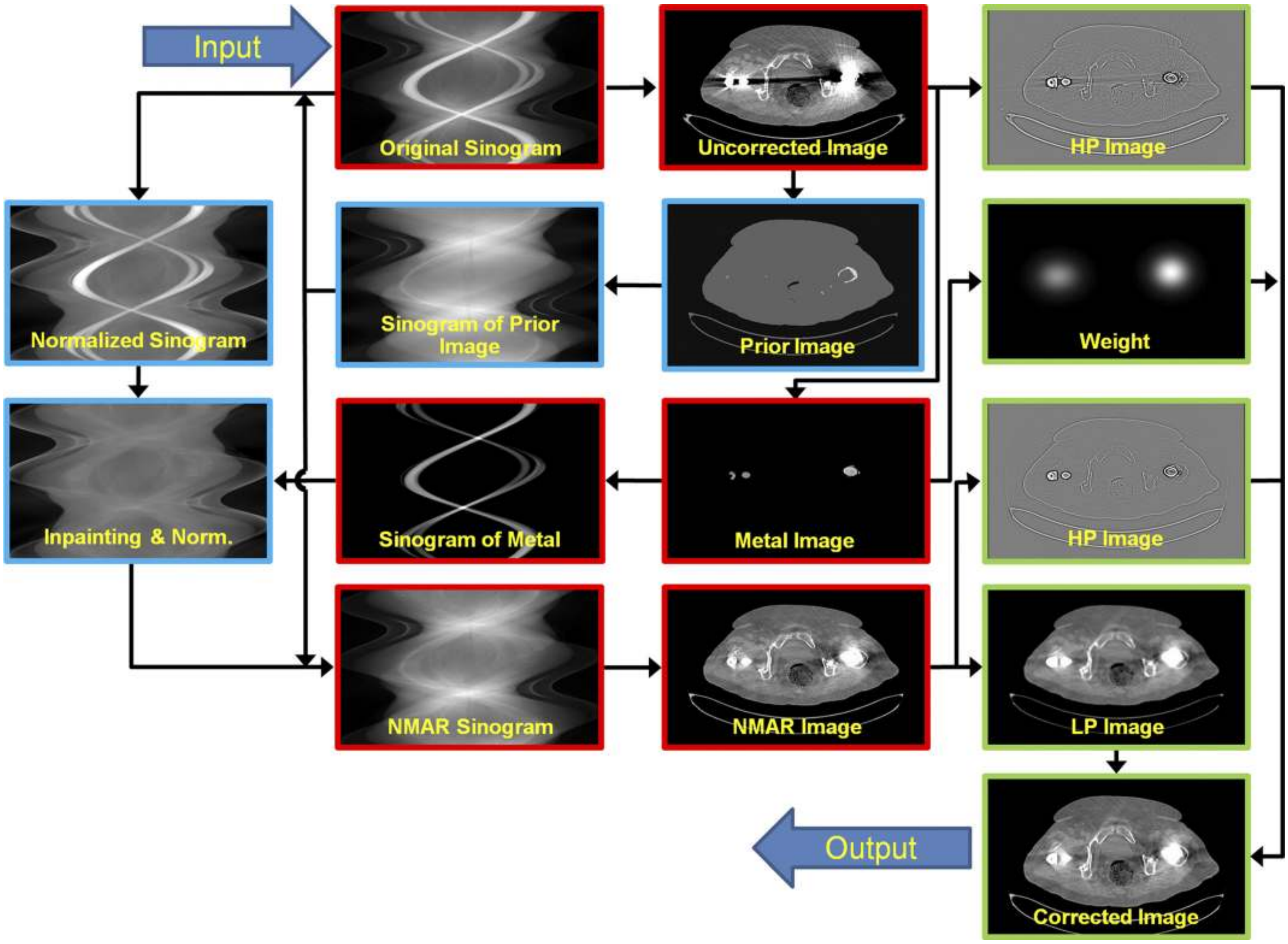

FIGURE 5. Dedicated metal artifact reduction algorithms either use simple inpainting (red), add prior information to normalize the data (red and blue), or additionally perform a frequency split to restore the noise texture (red, blue, and green). Figure 5 can be viewed online in color at www.investigativeradiology.com.

monochromatic DECT images may be a valuable alternative.[137,138] Flat-panel CT and cone-beam CT are considered alternative modalities in bone imaging, especially in the presence of metallic hardware, but MDCT has been demonstrated to provide sufficient image quality for postoperative imaging of internally fixated wrist fractures even without dedicated MAR algorithms.[139]

DUAL-ENERGY CT (DECT)

With the introduction of DSCT systems, DECT celebrated a revival from initial efforts of the 1980s.[140–143] Dual-energy CT makes use of the fact that attenuation can be decomposed into two linearly independent functions of energy, for example, into water and bone, into water and iodine, or, more physically, into photo and Compton effect. To allow for such decomposition, DECT performs a scan using two different x-ray spectra. Achieving optimal separation between the low- and the high-energy spectra is important in DECT because the quality of material decomposition depends on how different the two spectra are: the larger the difference between the two spectra, the smaller the noise level.

There are several implementations of DECT in routine clinical use (Fig. 6). Using two tubes, operating at different potentials, each tube's x-rays can be selectively prefiltered to minimize patient's dose and to improve spectral separation.[144] The additional prefilter on the high-energy x-ray source removes the undesired low-energy photons before they reach the patient. Another advantage of DSCT is that both x-ray tubes can be operated separately with individual tube currents and with individual tube current modulation curves (the low-energy thread requires larger modulation amplitudes than the high-energy thread).

Another implementation is based on fast tube voltage switching such that every other projection is performed at either the low- or the high-kilovolt value. This requires a dedicated x-ray tube and x-ray power generator, but selective prefiltering of the high-kilovolt x-rays is impossible. Owing to the finite rise and fall times of the tube voltage, the spectral separation is limited. In addition, the tube current cannot be switched owing to the temporal inertia of the filament and the required differences in mAs product for the low- and the high-kilovolt raw data can only partially be realized by setting the dwell times accordingly. An advantage of the kilovolt switching technology is that DECT can be provided in the full 50 cm field of measurement; and therefore, patient positioning is not as critical as with smaller field of measurements (26–35 cm), as it is the case with DSCT owing to space limitations.

Dual-layer, or sandwich, detectors are another alternative to realize full-field DECT. The first detector layer prefilters the x-rays; and thereby, the second layer sees the prefiltered x-rays that passed the first layer. Thus, the second layer measures the high-energy spectrum,

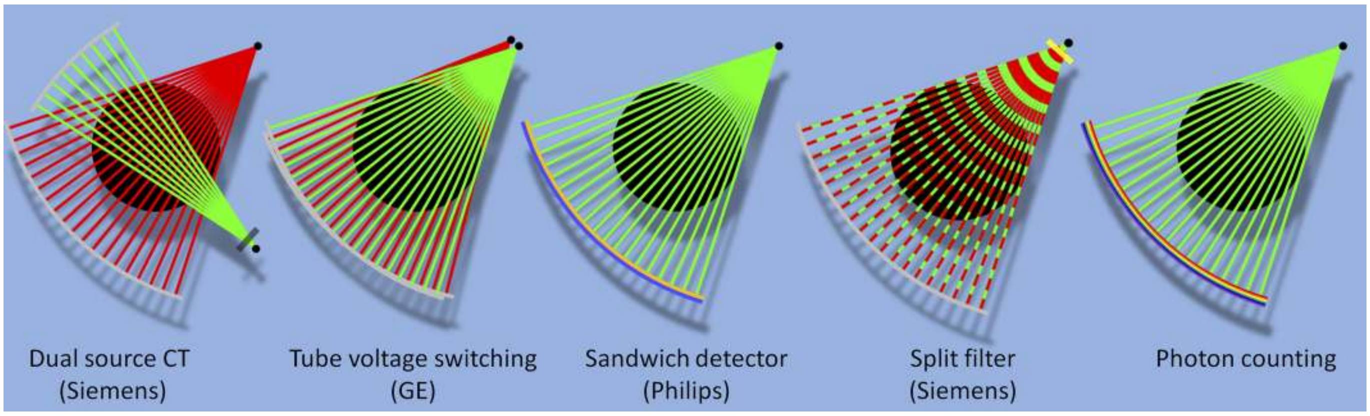

FIGURE 6. Today's dedicated DECT implementations and a potential future system with photon counting CT detectors. Not shown here is the option to do two subsequent scans with different tube voltages, as it is available for some mid-range CT systems. Figure 6 can be viewed online in color at www.investigativeradiology.com.

whereas the first layer mainly captures low-energy photons. The spectral separation cannot be adjusted with such systems, and separation will be limited, resulting in more noise. Sandwich detector DECT information provides geometrically fully consistent low- and high-energy data, which is not the case with the other implementations. Raw data-based DECT processing rather than image-based material decomposition is possible, and the user can retrospectively decide to use conventional single-energy CT or DECT information.

Recently, a split-filter technique was introduced in single-source CT, using a prefilter that is split along the *z*-direction such that (in the case of a 64-detector row system) the first 32 detector rows are prefiltered differently than the last 32 detector rows. A spiral pitch value below 0.5 needs to be chosen to ensure that each voxel is scanned with each of the two different spectra. The spectral separation achievable with such an approach is comparable to the sandwich detector and the tube voltage switching concepts. The system also provides spectral information throughout the full 50 cm field of measurement.

As previously mentioned, photon counting detectors are a highly promising technique for future diagnostic CT systems. Since these detectors can discriminate two or more energy windows, they are intrinsically suited for DECT applications. It should be noted that DSCT systems can potentially be equipped with one photon counting detector and one conventional energy integration detector, or with two photon counting detectors.

Typical clinical applications of DECT are virtual nonenhanced imaging,[145–147] automated bone removal,[148–151] urinary stone classification,[152–155] gout imaging,[80,156–162] metal artifact reduction,[135–138,163,164] cardiac[165–170] and pulmonary perfusion imaging,[171–183] and "monoenergetic" imaging (Figs. 7 and 8).[136,143,184–188] The very high CNR

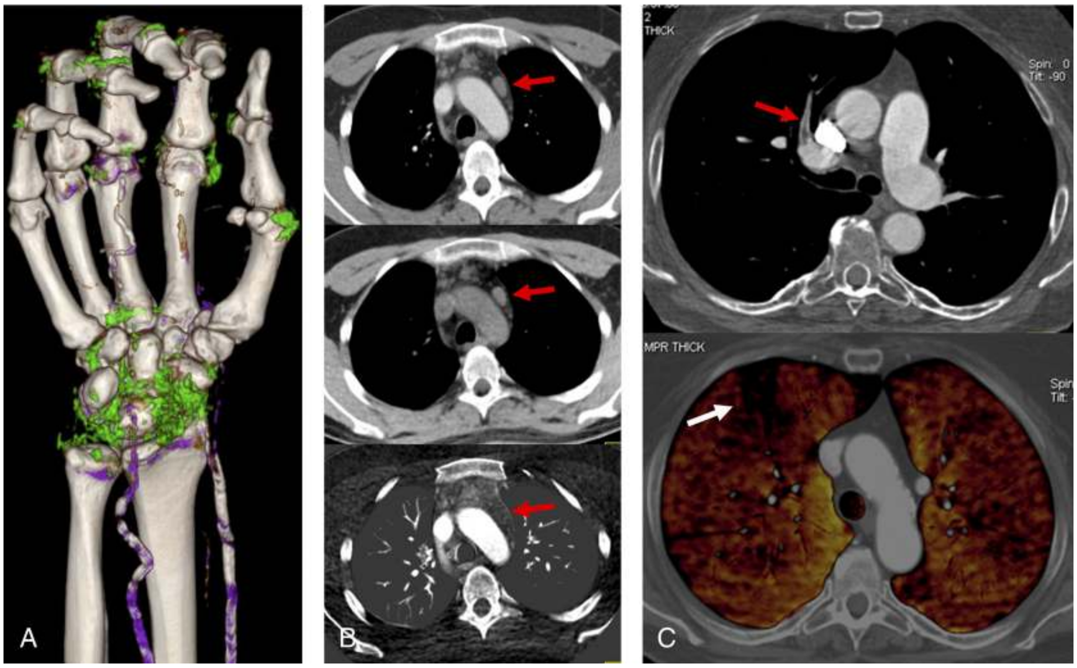

FIGURE 7. Typical examples of DECT. A, A 62-year-old patient with multiple urate deposits (coded in green color). Calcifications are coded in gray or purple (i.e, vessel walls). B, A 47-year-old patient after treatment of lymphoma. Persistent enlarged mediastinal lymph node (arrow) without contrast enhancement (upper row, 120-kV equivalent image; middle row, virtual nonenhanced image; bottom row, iodine map). C, A 68-year-old patient with acute pulmonary embolism. Embolic material (arrow) in CTA and the resulting perfusion defect (white arrow) in the color-coded iodine map (bottom row) can be visualized. Figure 7 can be viewed online in color at www.investigativeradiology.com.

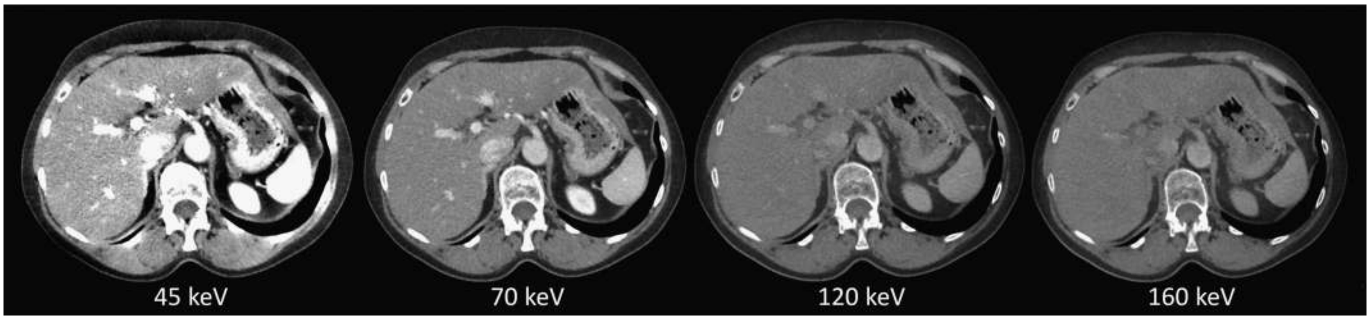

FIGURE 8. Energy dependence of iodine enhancement: monoenergetic imaging of a portal venous DECT realized with the split-filter technique. Very bright iodine contrast on 45-keV image (left) is gradually decreasing with increasing kiloelectron volt values.

in low–kiloelectron volt images can be used to either enhance the image contrast or to reduce the volume of CM while maintaining the CNR of a 120-kV image.

DYNAMIC CONTRAST-ENHANCED CT

Dynamic contrast-enhanced CT (DCE-CT) or perfusion CT (PCT) aims to assess the blood supply of tissue through analysis of the temporal enhancement pattern after CM injection. Rapid repetitive sampling of a specific tissue volume is performed while injecting a small volume of CM at a high injection speed.[189–191] The concentration of iodine within blood vessels and tissue is linearly proportional to the increase in attenuation; and therefore, the time-attenuation curve (TAC) can be analyzed for each voxel using a kinetic model. Hemodynamic parameters can be calculated from the distribution of CM in the intravascular and extracellular compartments: Mean transit time (MTT, time the tracer needs to travel through tissue vasculature), regional blood flow (BF, blood flow per tissue unit), blood volume (BV, fraction of blood per tissue unit), extraction fraction (E, fraction of tracer that is extracted to extracellular space during first pass), flow extraction product or permeability (FE or K_{trans}, tracer flow from intravascular into extracellular space per tissue unit), permeability surface area product (PS). Different techniques have been developed to postprocess the DCE data: for brain perfusion imaging, the deconvolution and maximum-slope approach; for tumor perfusion imaging, the deconvolution, Patlak method (BV, FE), and maximum slope. The limited volume coverage of older CT systems restricted the clinical use of DCE-CT predominantly to stroke imaging,[192–198] but the advent of high-frequency spiral techniques (variable pitch spiral) and very large area detectors[191,199–202] facilitated DCE-CT in whole-organ imaging. Lesion characterization, response prediction, and assessment are in the center of interest in oncologic imaging and extend the insights in tumor biology and treatment.[143,203–215] Dynamic contrast-enhanced CT is also used to assess physiological and pathophysiological organ function, for example, kidney function[190,199,212,216–225] or myocardial perfusion.[200,202,226–233] Myocardial perfusion imaging faces further challenges owing to the fast heart motion and the increased susceptibility to

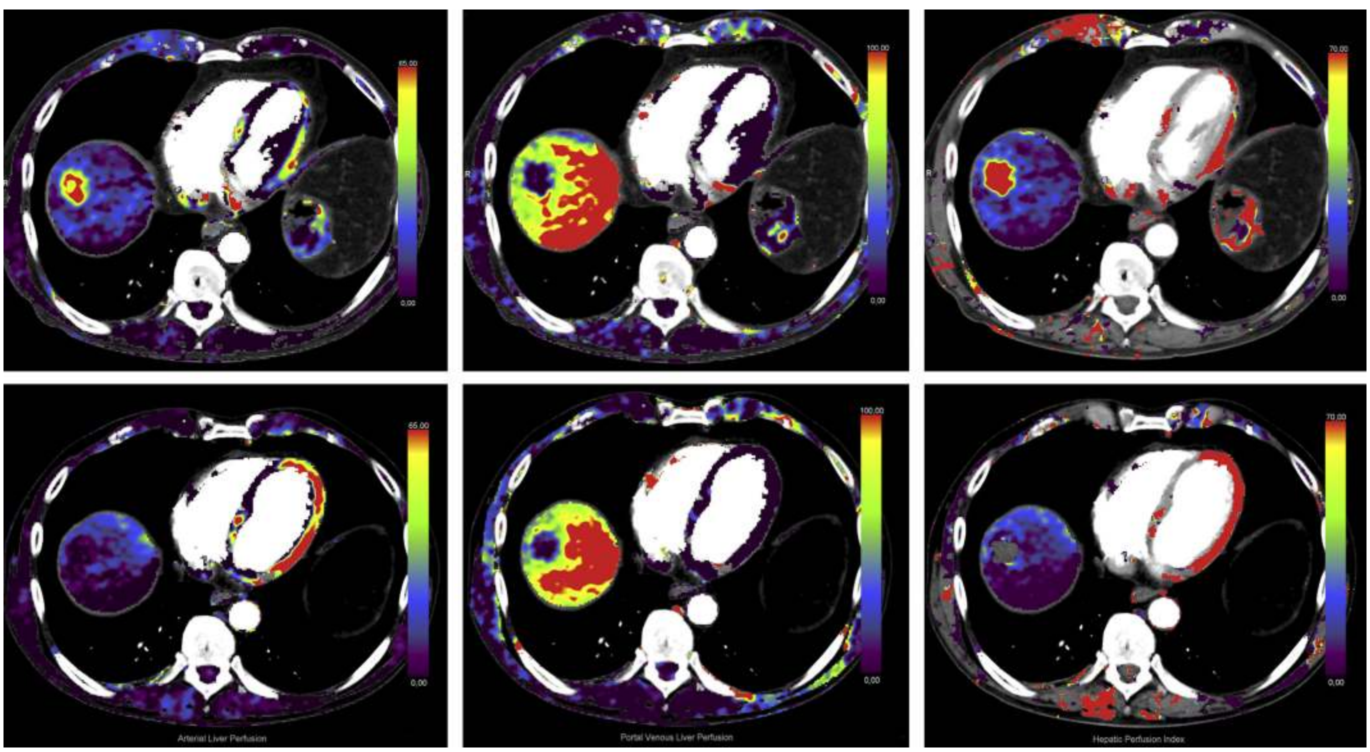

FIGURE 9. A 60-year-old man with HCC: upper row, before; bottom row, after therapy (TACE). Parameter maps demonstrate complete devascularization of the tumor after chemoembolization. From left to right, arterial liver perfusion, portal-venous liver perfusion, and hepatic perfusion index. Figure 9 can be viewed online in color at www.investigativeradiology.com.

artifacts. The heart motion can be compensated to some extent by synchronizing the acquisition with the ECG signal. However, this introduces strong limitations in the temporal resolution, which is often too low to follow the tracer dynamics, which are typically much faster in the heart than in other organs. For these reasons, further assumptions and efforts have to be made in the modeling step. Furthermore, beam hardening artifacts due to iodine or bone tend to change the CT values in the surrounding tissue.[234] Partial scan reconstruction used in cardiac CT to maximize the temporal resolution introduces CT value variations as a function of the start angle of the partial scan segment.[235] Dedicated algorithms to compensate for these inconsistencies need to be applied.[236–238] Chest perfusion imaging is discussed in more detail in Goo et al.[239]

Hepatic perfusion imaging[191,206,240–245] has gained special attention in a variety of clinical scenarios (tumor classification and grading, and therapy response assessment). Unlike most other organ systems, the liver (and lung) has a dual blood supply, necessitating adapted data postprocessing and modeling (Fig. 9).

Dynamic contrast-enhanced CT implies repetitive scanning of the same organ while the CM is passing the tissue of interest. Total scan times of 40 seconds up to several minutes have been proposed. This has two implications: patients are not able to hold their breath for such a long time, and radiation exposure increases. Registration and motion correction algorithms are used to cope with the first problem, low-kilovolt scanning, and all other means of dose reduction as previously described need to be considered for the second. Specific data such as the extracellular volume fraction and arterial enhancement fraction, which significantly correlate with hepatic fibrosis and cirrhosis, can also be determined with conventional multiphasic liver CT,[246,247] simplifying the process of data acquisition at moderate radiation exposure. As indicated in the previous section, DECT can provide iodine maps, which are often regarded as perfusion images, but it is important to recognize that DECT images do not reflect dynamic information but rather the iodine distribution inside an organ or a lesion at a specific time point.

CONCLUSIONS

Although repeatedly declared dead, CT has celebrated an amazing comeback within the past decade. Computed tomography is the workhorse in daily practice, spreading into new applications like cardiac and quantitative imaging, and shifting indications from radiography to CT with major clinical impact. Progress is seen with increased spatial resolution, faster scan speed, lower CM and radiation dose, and morphological and functional information. In 2014, approximately 81.2 million CT procedures were performed in the United States,[248] which on the one hand highlights the role of CT but on the other hand emphasizes the need for an intelligent dose management.

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