2 The Technology

2.1 Scanner Characteristics

2.1.1 Anatomical Coverage

Multidetector computed tomography (MDCT) expanded the volume coverage per gantry rotation and revolutionized the imaging of many body regions. The most dramatic advances in MDCT technology profited from the increase in the number of detector rows. In general, the wider the detector, the larger the anatomical coverage. The dimension of the detector can be calculated by multiplying the width of a single detector's element by the number of elements. Currently, the widest anatomical coverage is 16 cm and it is obtained using a 320-slice detector with 0.5-mm slice thicknesses (320 × 0.5 mm = 16 cm).

The main advantage of wider volume coverage is faster MDCT acquisition, which is extremely important to decrease motion artifacts. A second advantage is the capability to evaluate whole-organ perfusion (e.g., cardiac stress imaging, whole liver and brain acquisition) without table movement, thus reducing related artifacts. The possibility to acquire large body segments also enables fluoroscopic examinations of the upper gastrointestinal tract and visualization of joint instability with dynamic imaging.

2.1.2 Temporal Resolution

Temporal resolution refers to the time needed to acquire a complete data set for image reconstruction. Superior temporal resolution improves the ability to freeze moving structures such as the heart. The temporal resolution of a single X-ray MDCT tube and detector system is one half of the gantry rotation time, because image reconstruction requires approximately 180°

plus the fan angle. The rapid evolution of multidetector technology has improved gantry rotation speeds, reducing temporal resolution. When the MDCT standard reached 64 slices per gantry rotation, the gantry rotation time was 360 ms but it is now as low as 270 ms.

Temporal resolution also can be improved by the introduction of a second X-ray MDCT source with an independent detector system. Dual-source MDCT features two X-ray MDCT sources and detector systems separated from one another by 90°. Consequently, when both MDCT tubes are used, the temporal resolution is halved, from 165 ms to 83 ms, because only a quarter gantry rotation is needed for reconstruction. This implies, for example, the possibility to perform cardiac studies in patients with a rapid heart rate, without the need to administer beta-blockers but still obtaining high-quality images.

2.1.3 Spatial Resolution

The spatial resolution of a MDCT system refers to its ability to separate two structures and is determined by measuring the ability of the system to separate line pairs. Voxel size has become the surrogate parameter for spatial resolution; it is determined by the field of view (FOV), image matrix, detector efficiency, number of projections (sampling frequency), and section thickness. Even before the advent of 64-row MD-MDCT, most MDCT image reconstruction algorithms included isotropic voxels, which means that the x, y, and z dimensions are cubic with equal lengths along each side. Isotropic voxels enhance image post-processing, achieving excellent image quality in any plane and the generation of excellent volumetric reconstructions. High spatial resolution is critical for the analysis of small structures such as the coronary arteries or implanted stents. Currently, a 0.35-mm spatial resolution is achievable with the latest-generation MDCT scanners.

The signal to noise ratio (SNR) is another important parameter intimately related to spatial resolution. SNR is defined as the amount of signal, or information, that can be used for interpretation divided by the image noise. There is no information in noise; it is akin to an imperfect canvas on which the image data, or signal, is painted. As the MDCT slice thickness decreases, so does the amount of tissue per slice and hence the signal. However, a significantly higher SNR data set, i.e., less noisy images, can be obtained when images are reconstructed at double the slice thickness, for example 0.8 mm instead of 0.4 mm.

2.1.4 Contrast Resolution

Recent advancements in the composition of detector materials have improved MDCT contrast resolution, which is the ability to distinguish between differences in intensity within an image. Contrast resolution in MDCT can be changed with the application of different filters. Less filtering makes the beam softer and yields a higher-contrast image but it is also associated with higher radiation exposure because soft radiation is more readily absorbed by biologic tissue. Conversely, more filtering yields lower-contrast images while reducing radiation exposure. The effect may be negligible on unenhanced MDCT scans but is clearly apparent on scans obtained after the administration of an iodine-based contrast agent, mainly because of the differences in the radiation absorption of different energies by iodine.

2.2 Systems for Reducing Dose Exposure

2.2.1 Limiting z-Axis Length

In everyday practice, scan coverage is set up visually on the scout-topogram. Since the total radiation dose delivered is directly proportional to z-axis coverage, the accurate adjustment of scan length is important for optimization of the dose-length product (DLP).

2.2.2 Tube-Amperage and Tube-Voltage Adjustments

Adjustment of the tube current and/or the voltage applied to the X-ray tube reduces the number and/or average energy of the photons generated, respectively. Furthermore, both directly affect image SNR. The radiation dose is approximately proportional to the number of mAs and is adjustable, thereby allowing amperage values to be customized according to body mass. Failure to adjust this parameter downwards for thin patients will result in unnecessary radiation. It is important to reduce mAs as much as possible while ensuring a high enough SNR to obtain diagnostic-quality images.

Another approach to reduce the dose delivered to the patient is to reduce the tube voltage. Tube kVp affects both peak photon energy and image contrast. Tube voltage has a more dramatic effect on radiation dose, which is exponentially proportional to kVp. However, the use of photons with a lower average energy leads to an increase in the attenuation values of elements of higher atomic number and greater absorption coefficients because of the increased interaction resulting from the photoelectric effect; the greater disadvantage of this approach is a reduction in diagnostic accuracy caused by increased beam-hardening artifacts in the calcified plaque as well as a reduction of the SNR, even if the latter is compensated by a slight increment in tube current.

2.2.3 Automatic Modulation of Tube Current

Another option is based on anatomy-adapted tube-current modulation. Thus, according to the individual characteristics of the patient, the X-ray tube current can be adjusted using attenuation values quantified by the patient's initial scout image. The pixel noise in a MDCT image is largely attributable to the projections in which the greatest attenuation occurs; the intensity of the radiation can be reduced in projections with less attenuation. Thus, the tube current will be reduced in thin patients, or rather in those with reduced anteroposterior thickness (*xy* plane), and greater in obese patients, in order to obtain optimal image quality.

2.2.4 Iterative Reconstruction

Iterative reconstruction techniques have demonstrated the potential to improve image quality and to reduce the radiation dose in MDCT relative to the currently used filtered back-projection techniques. The most noticeable benefit of iterative reconstruction is that it is able to incorporate into the reconstruction process a physical model of the MDCT system that can accurately characterize the data acquisition process, including noise, beam hardening, and scatter. This ability dramatically improves image quality, especially in the case of low-dose MDCT scans, in which the propagation of non-ideal data during image reconstruction becomes more significant than in routine MDCT scanning. Iterative reconstruction is also superior to filtered back-projection in handling insufficient data. Recent advances in iterative reconstruction allow a significant reduction in the number of required projection views while still producing acceptable image quality. Iterative reconstruction techniques thus have the potential to substantially reduce the radiation dose in MDCT. With computational power growing quickly, the clinical implementation of iterative reconstruction algorithms is within delivery.

2.3 Multi-Energy

Multi-energy computed tomography (MECT), i.e., the acquisition of data sets at different photon spectra in a single MDCT acquisition, can provide information on the material composition of the tissues based on differences in photon absorption. This enables materials of similar density but different elemental composition to be distinguished from one another, especially materials with large atomic numbers such as iodine.

In particular, the use of dual-energy post-processing software, based on three-material decomposition principles (soft tissue, fat, and iodine), allows the contrast agent to be distinguished and isolated from other materials. The possibility to accurately recognize iodine distribution enables the generation of a virtual unenhanced or virtual angiographic data set. The generation of virtual unenhanced images may obviate the routine need for a pre-contrast scan, thus decreasing patient radiation exposure especially in multiphasic acquisitions or repetitive MDCT examinations. Indeed, as reported in the literature, this means a dose reduction of about 30% .

The ability to accurately discriminate parenchymal iodine distribution using a color-coded image implies improved quantification of both subtle tumor vascularization and the response to anti-angiogenic therapy.

Liver MECT may permit the rapid detection and quantification of hepatic iron overload in thalassemia patients.

Finally, the advantage of different energy acquisitions has been investigated, with the results showing that hypervascular lesions are better visualized and show increased conspicuity at 80 Kv than on a standard 120-Kv scan.