Introduction

The quest for non-invasive imaging of cardiac maladies began with conventional radiography and continues today. Echocardiography has established itself as the screening and diagnostic modality of choice for a variety of cardiac ailments. Since the early 1990s, cardiac magnetic resonance imaging (MRI) has also become an imaging standard of reference for a variety of cardiac pathologies, including cardiac tumors, myocarditis, cardiomyopathies, viability, complex congenital heart diseases, and valvular heart diseases. However, up until the development of multi-detector computed tomography (MDCT) scanners, the imaging of coronary artery diseases with a noninvasive imaging test was a largely elusive approach; instead, invasive conventional coronary catheterization remained and, to date, firmly remains, the imaging modality for most coronary artery diseases.

With the introduction of MDCT scanning into clinical practice, perhaps no other CT application has grown as much as coronary CT angiography (CTA) [1]. The speed and resolution of modern MDCT scanners has made it possible to obtain “acceptable quality” studies for those subjects with slow and regular heart rates. The development of multi-detector computed tomography (MDCT) scanners, the imaging of coronary artery diseases with a noninvasive imaging test was a largely elusive approach; instead, invasive conventional coronary catheterization remained and, to date, firmly remains, the imaging modality for most coronary artery diseases.

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Faster Imaging

The earliest CT “step-and-shoot” conventional scanners were encumbered by the wires attached directly to the X-ray tube and detector array panel, with the consequent need to rewind. These scanners also had smaller z-axis coverage per gantry rotation. The slip-ring technology in helical CT scanners allowed engineers to attach these wires to a ring and then mount the tube and the detector array panels onto the ring, thus avoiding the need for rewinding. Thus, helical CT scanners allowed rapid continuous acquisition, whereas the multi-detector-row helical CT scanners made single-detector-row helical scanners truly volumetric, with extended z-axis coverage. Sub-second X-ray tube rotation times of MDCT scanners allowed the rapid acquisition of image data with multiple submillimeter detector elements arranged in multiple rows, enabling acquisition of a near-isotropic image data set.

The 4-slice MDCT scanners allowed subjects with slow and steady heart rates to be imaged with sufficient-quality images of the coronary arteries and reasonable sensitivities and specificities for proximal coronary segments [4]. However, these 4-slice scanners had several limitations. Due to the relatively thin beam collimation and use of small overlapping pitch, image quality in patients with rapid or irregular heart rates was limited, as was the evaluation of more distal segments of coronary arteries. Likewise, these factors also mandated acquisition of a large number of rotations, which were associated with greater slab artifacts and longer breath-hold durations. Most 4-slice scanners were also limited by poor radiation-dose efficiency compared with their predecessors, the conventional non-helical and single-slice-single-detector row helical scanners. Introduction of 8-, 16-, 32-, and 64-slice MDCT scanners with improved detector geometry, wider beam collimation, and efficient online pre-patient beam collimations led to substantially better radiation dose efficiency and speed of acquisition for coronary imaging than obtained with the 4-slice scanners.
Based on MRI experience, it has become clear that temporal resolution, for at least the functional part of cardiac analysis, needs to be less than 50 ms. Depending on the vendor type, 64-slice MDCT scanners have beam collimations of 2–4 cm and a single-segment temporal resolution of 165–200 ms. Although with multi-segment reconstructions it is possible to reduce the 165- to 200-ms duration to 83–100 ms for two-segment reconstructions and to around 50 s for four-segment reconstructions, the use of such techniques requires that the patient have a regular heart rate and that data are averaged over more than one cardiac cycle. Furthermore, such multi-segment reconstruction of a cardiac data set requires the use of extremely small, overlapping pitch, which increases radiation dose and scan duration. The latter, in turn, requires a slower contrast medium injection rate and volume as well as longer breath-holds.

Also, 2- to 4-cm beam collimation requires multiple gantry rotations over several heartbeats to image the entire heart, which can lead to stair-step and slab artifacts.

Another limitation of current single-source MDCT scanners is in the imaging of large patients [5]. Increased image noise due to large body habitus can substantially drop the contrast-to-noise ratio and lead to suboptimal studies. Although the power of modern X-ray tubes is better, allowing use of higher tube current, as for scanning of other body regions, coronary imaging with these scanners may not provide the required information in very large or obese patients [5, 6].

Evaluation of coronary stents and calcified plaques with single-source MDCT scanners is also hampered due to blooming artifacts, which can lead to overestimation of the degree of vessel lumen stenosis or even to a non-evaluable study [7, 8]. Likewise, evaluation of the distal coronary arteries and collateral circulation is also affected by the limited spatial resolution of the currently available CT scanners.

Efforts to improve the temporal resolution of MDCT scanners could have involved improving the X-ray tube rotation speed. However, as the force on the gantry components increases in proportion to the square of the angular velocity, it was difficult to improve the gantry rotation speed [1]. Therefore, the concept of using two X-ray tubes and detector assemblies with the same gantry rotation time (330 ms) in order to improve temporal resolution, was pursued [2, 3]. In order to reconstruct the images, data from about 180° or a half-gantry rotation (plus the fan angle of the X-ray beam) are needed. Thus, with single-source CT scanners, one-segment reconstruction needs at least one-half of the gantry rotation time. However, for dual-source CT scanners, a single-segment reconstruction of at least one-half gantry rotation can be obtained by fusing data from quarter rotations of two X-ray tubes, thereby improving the single-segment reconstruction time to about 83 ms. With dual-segment reconstruction, dual-source CT has a temporal resolution of 60 ms (with a minimum of 42 ms) [2]. It is interesting to note that multiple X-ray source design is not new to CT scanning. Almost three decades earlier, a multiple X-ray source CT scanner, the high temporal resolution cylindrical scanning computerized tomographic system (Dynamic Spatial Reconstructor), was developed to image the heart. It consisted of 14 X-ray tubes mounted on a single semicircular arch [9, 10]. Unfortunately, use of this scanner was limited to research purposes due to the poor signal-to-noise ratio and computational power for image reconstructions required for human use.

While the DS-MDCT scanner attempts to improve temporal resolution, its beam collimation and spatial resolution are identical to those of its corresponding predecessor, the single-source 64-slice MDCT scanner.

**Practical Physical Aspects of the Dual-Source Scanner**

A dual source scanner is basically a 64-slice MDCT scanner with two X-ray sources and two detector arrays (Table 1).

**X-Ray Tubes**

Dual-source MDCT scanners have two X-ray tubes, each with a power equal to a single-source 64-slice MDCT scanner. Thus, the combined power of the X-ray tubes in dual-source scanners is two-fold greater than that of a single-source scanner (160 kW compared with 80 kW) [3]. These X-ray tubes are mounted orthogonally from each other on the same gantry assembly (Fig. 1). They can be operated at similar or different kilovoltage (kVp) and milliamperage (mA). For cardiac imaging, the best temporal resolution can only be obtained at similar kVp for the two tubes. With different kVp or dual energy, a single-segment temporal resolution of 165 ms is achievable, which is identical to that of a 64-slice MDCT scanner. Like its predecessor 64-slice single-source CT scanner, both X-ray tubes have z-flying focal spots, which enable the acquisition of 64 slices from a 32-slice detector assembly. Also, to reduce the radiation dose, an additional targeted field-of-view (FOV) cardiac beamshaping or bowtie filter is used. McCollough et al. showed that the addition of this filter reduces radiation dose by 17% compared with the body beamshaping filter only [11]. It is also important to re-