Introduction

The heart can be visualized in gross form on any standard chest computed tomography (CT), but for detailed evaluation of cardiac anatomy exacting specifications of scanner hardware and software must be in place because: the heart is continually in motion, the structures of interest (i.e., coronary arteries) are small, and their curvilinear course requires multiplanar reformatting for analysis.

For these specifications to be met, the scanner hardware must be robust, the scanner must be fitted with ECG-monitoring equipment, and cardiac-specific software and image reconstruction servers must be installed.

Elements of Scanner Design

The general design of a CT scanner equipped for cardiac imaging is the same as that of any modern scanner: an X-ray tube mounted on a rotating gantry opposite a detector array generates X-rays which are emitted across the bore, are attenuated as they pass through the patient's chest, and are scintillated by the detectors as the gantry rotates around the patient obtaining attenuation data from multiple view angles. These data are transmitted to an image reconstruction server which then computes the axial images. To perform cardiac CT, however, a number of additional capabilities are required, [1, 2] namely, high spatial resolution, high temporal resolution, and fast scan speed, while delivering an acceptable radiation exposure.

Special Design Considerations for Cardiac CT

Spatial Resolution

Spatial resolution is the product of multiple variables including detector characteristics and collimation, sampling rate, and reconstruction methods. State of the art scanners use thin 0.5–0.625 mm collimation. This reduces volume averaging thus increasing image detail of small objects, and allows reconstruction of isotropic datasets, thereby preserving z-axis resolution and permitting multiplanar reformatting necessary for analysis of cardiac anatomy.

GE Healthcare has focused their system development on spatial resolution, developing a new detector scintillation material with a fast decay time (30 ns), minimal afterglow (one-quarter that of conventional Gd2O2S crystals), and increased efficiency [3]. This new detector is one element of a multipronged approach to increasing image resolution, which also includes new electronics that permit increased sampling,[4] and a new iterative reconstruction algorithm [5, 6] (Figure 2.1). These innovations increase spatial resolution by 3–4.5 line pairs per centimeter (lp/cm), which is substantial given that current resolution of systems is approximately 12–15 lp/cm. The enhanced resolution is particularly useful for imaging stented and/or calcified coronary arteries.

Temporal Resolution

In order to reconstruct an axial image, attenuation data from multiple view angles that encompass half a rotation around the patient are required. The amount of time needed to obtain these data is the nominal, heart-rate independent temporal resolution of the scanner. Hence, a gantry that requires 330 ms to make one complete rotation has a baseline temporal resolution of 165 ms. This is analogous to shutter speed: just as shutter speed must be sufficiently fast to capture motion-free images of a moving subject, temporal resolution must be sufficiently fast to capture the data needed for image reconstruction within the relatively brief period of cardiac diastasis. If temporal resolution is insufficient, motion artifact occurs (Figure 2.2). A temporal resolution of 165 ms is relatively slow for cardiac imaging, so systems overcome this by various means: faster gantry rotation, multicycle reconstruction, and a second X-ray tube.
Currently, the fastest gantry on the market resides in the Philips iCT, which has a rotation time of 270 ms using an innovative frictionless design in which the gantry is actually suspended in air [7]. As this design has only recently been released, it is likely that future iterations will allow even faster rotation.

Multicycle reconstruction combines data from multiple contiguous heart beats to complete the half-scan of views (Figure 2.3). This technique is heart-rate dependent: it can only be used within certain ranges of heart rates, since the cardiac cycle and gantry rotation must be asynchronous.

An approach adopted by Siemens in 2006 uses two X-ray tubes mounted on the gantry at 90° angles; therefore, only a quarter turn of the gantry is required to collect the 180° of attenuation data (Figure 2.4). Hence, a dual-source system with a gantry rotation time of 330 ms has a baseline central temporal resolution of approximately 85 ms. This obviates the need for multicycle reconstruction, making the system less susceptible to variation in heart rate, and providing high heart rate independent temporal resolution. Initial reports demonstrated very good image quality, [8] and, in a comparison of 64 MDCT and DSCT scans, the dual-source
system produced better image quality at higher heart rates [9]. Image quality is still improved, however, by reducing heart rate before scanning. Siemens recently announced a new scanner with an even faster (280 ms) gantry which is capable of a central temporal resolution of 75 ms [10].

**Detector Coverage and Scan Speed**

Since the advent of multislice CT, manufacturers have increased z-axis coverage by the addition of more rows of detectors. By increasing coverage, scan times have been reduced to just a few seconds, and even to single-beat scanning. This has been achieved by Toshiba’s Aquilion One scanner, which currently has the largest z-axis coverage [11]. This scanner has 320 rows of detectors, with 0.5 mm collimation, 16 cm of coverage, and can perform an axial 1-beat acquisition of the entire heart when the rate is well controlled.

The Philips iCT is a 128-row scanner with 8 cm of coverage, which allows axial acquisition of the entire heart in 2 beats [7]. GE offers a 64-row scanner with 4 cm of coverage. The current Siemens system is a 32-row scanner, with approximately 2 cm of coverage, but Siemens has recently announced a new 64-row scanner offering approximately 3.8 cm of coverage.

**Radiation Exposure**

Traditional cardiac CT uses a helical scan mode and very low pitch to perform retrospective ECG-gated reconstructions. This delivers an undesirably high radiation exposure, so various methods have been designed to reduce radiation exposure.

One of the first was ECG-based tube current modulation, which fluctuates tube current in sync with the heart rate, maintaining high tube current during diastasis and providing best image quality in that phase, then lowering tube current during other phases when high-resolution detail is not required (Figure 2.5). This results in a dose savings of approximately 40%.

A great advance in dose reduction was made with the advent of prospective, ECG-triggered, axial cardiac CT (Figure 2.6). By keeping the X-ray tube off during most of the scan, and only turning it on during diastasis, as triggered by the ECG, radiation exposure is dramatically reduced (Figure 2.7). GE first published results using this method [12] with good clinical results and radiation doses of 1–2 mSv, [13] and this approach is now offered by all manufacturers. Using a 64-slice system, the entire heart can be covered in 3–4 acquisitions; however this leaves the system vulnerable to fluctuations in heart rate and rhythm during the scan. Larger coverage mitigates this vulnerability by reducing the number of beats.
required for data acquisition. However, larger coverage lends to higher radiation doses, as the detectors are less efficient when having to expose a wider detector array. Hence, Philips’ 128-row iCT system can cover the entire heart in 2 beats, with a reported radiation exposure of 3–5 mSv [7]. Toshiba’s 320-row Aquilion One scanner can cover the entire heart in one beat, with a reported radiation exposure of approximately 6 mSv. However, lower radiation doses and dependable coronary scanning using 1-beat acquisition requires heart rate reduction. At higher heart rates (>65 bpm), optimal image quality requires a 2- or 3-beat acquisition (and use of multi-cycle recon) which increases radiation exposure to approximately 13 (2-beat) to 19 (3-beat) mSv [14, 15]. Acquiring with a wide exposure window (“padding”) can ensure the capture...
Figure 2.8. Prospective CTA phase selection and padding. Acquisition phase and padding depend on heart rate: For low heart rates (a), target late diastole (75% phase) for prospective acquisition, but if heart rate is high (b) (>75 bpm), target end-systole (40% phase) which is more likely to produce motion-free images. By restricting acquisition to the minimum exposure window (dark gray), radiation dose is minimized, but ability to reconstruct adjacent cardiac phases is obliterated. Widening the acquisition window (“padding,” light gray shading) allows reconstruction of a small number of additional phases, which can help interpretation of motion artifact, but increases the radiation dose. (Reproduced with the kind permission from Springer Science + Business Media from Weigold et al [7]).

Figure 2.9. Method of high-pitch coronary CTA. Single source helical CT requires a pitch ≤1.5; a faster pitch results in gaps in the data (left panel). A dual-source system can scan at a higher pitch (up to 3.2), using the second X-ray source-detector system to fill in what would otherwise be gaps in the data. Since the table speed is so high, the entire heart can be imaged in a fraction of a second, from data derived from a single heartbeat (right panel). If the image acquisition is triggered appropriately to synchronize acquisition with diastasis, and the heart rate is sufficiently low, motion-free images can be obtained with a very low radiation dose. (Reproduced with permission from Elsevier from Achenbach et al [18]).

of motion-free data, but increases radiation exposure (Figure 2.8). A narrower exposure window can be used with preservation of image quality if heart rate is controlled [16] and reduces radiation exposure.

Siemens has recently introduced a new innovative scanning method using a high pitch helical mode which takes advantage of the dual-source design [17, 18] (Figure 2.9). The high pitch (3.2–3.4) would normally produce gaps in the attenuation data using a single-source system, but, in a dual-source system, these are compensated for by data gathered from the second detector. Scan time is less than 1 s, with a radiation exposure of approximately 2 mSv. Initial studies performed on the first-generation dual-source system proved the feasibility of the method, though the authors noted that these first-generation systems are not suitable for this high pitch technique. High pitch scanning with their new scanner, with faster gantry rotation (280 ms) and larger coverage (64 rows), has been reported to produce very good coronary image quality with radiation exposure of approximately 1 mSv [10]. Of note, in this initial investigational phase, low heart rates (<60 bpm) are absolutely required, as the entire data collection takes place...
within a 250–270 ms acquisition window of one heart beat; hence diastasis must be at least this long to provide motion free data acquisition, and therefore the requirement for a very low heart rate.

The previously described new detector and reconstruction system offered by GE aims to lower radiation exposure by preserving image quality while scanning with less radiation (<1 mSv), because images are of higher resolution and lower noise than would otherwise be achievable with a low-exposure scan.

### Future Directions

Future scanner designs are closely guarded industry secrets, but some concepts have been openly discussed. Flat-panel volume CT systems replace detector rows with a large area detector, and provide coverage of the entire heart in one axial acquisition and extremely high spatial resolution of up to 26 lp/cm, comparable to invasive angiography [19] (Figure 2.10). However, the contrast resolution is inferior to that of multidetector CT, a high radiation exposure is needed to achieve a sufficient contrast-to-noise ratio, and scintillation times are slow and temporal resolution insufficient for cardiac imaging.

Dual energy scanning is being developed by all vendors by various means, including rapid fluctuation of tube energy, stacked detectors, or the dual-tube design in which each tube emits photons of different energy levels. Dual energy CT could take the field to a new level of diagnostic power if it can be used for refined tissue characterization, such as differentiating plaque characteristics. This is difficult to do in the coronary arteries, and initial studies have not yielded any breakthroughs, but at the least it has the potential to enhance coronary lumen visualization in heavily calcified vessels by differentiating calcium and iodine [20] (Figure 2.11). Dual-energy CT may also yield new applications for noncoronary imaging, such as imaging of myocardial perfusion [21].

### Conclusions

Since the turn of the twenty-first century, there has been an explosion in technological development of cardiac CT systems, with concomitant gain in reliability and accuracy, especially of coronary imaging. The theme has been one of progressively improved spatial and temporal resolution, reduced scan time, and, most recently, a focus on driving down radiation exposure. In appropriately selected patients, using careful technique, it is easily achievable to perform a cardiac CT using less radiation exposure than that of a standard chest CT or an invasive coronary angiogram. The goal of future systems will be to make this more widely achievable in a larger group of patients without requiring patient selection or stringent patient prep. Given the history of rapid technological advancement, we can expect to see this goal achieved in the near future.
Figure 2.11. Dual energy CT. Dual energy CT can be used to characterize tissue: vessel cross-sectional images derived from high- (140 keV) and low-energy (90 keV) attenuation profiles (a, b) demonstrate the influence of photon energy on photoattenuation. The contrast-filled lumen (arrow) exerts greater photoattenuation on low-energy photons, and hence appears brighter in (b), while calcification (asterisk) appears high-density in both images. In the subtracted image (c), dense calcification has been “removed.” By adding in the low-energy (90 keV) attenuation data to this subtracted image, depiction of the lumen edge is enhanced (because of reduced contrast blooming), and visualization of a small side branch (arrowhead) is improved (d). (Reproduced with permission from Wolters Kluwer from Boll et al.[23]).

References

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