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State-of-the-art in CT hardware and scan modes for cardiovascular CT

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Abstract

Multidetector row computed tomography (CT) allows noninvasive anatomic and functional imaging of the heart, great vessels, and the coronary arteries. In recent years, there have been several advances in CT hardware, which have expanded the clinical utility of CT for cardiovascular imaging; such advances are ongoing. This review article from the Society of Cardiovascular Computed Tomography (SCCT) Basic and Emerging Sciences and Technology (BEST) Working Group summarizes the technical aspects of current state-of-the-art CT hardware and describes the scan modes this hardware supports for cardiovascular CT imaging.

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Keywords

Multidetector row CT; CT hardware; scan modes; cardiovascular CT; coronary CT angiography; helical; axial; high-pitch; ECG-gating; ECG-triggering

Introduction

Since the late 1990s, research has shown that computed tomography (CT) imaging allows for noninvasive imaging of the heart and great vessels, especially the coronary arteries. Imaging the coronary arteries is very challenging because of their small dimensions and motion. Cardiovascular CT requires high temporal resolution to “freeze” cardiac motion and has therefore benefited from steadily increasing gantry rotation speeds. Data acquisition and image reconstruction fully synchronized with the electrocardiogram (ECG) signal are also necessary to obtain image datasets from the desired cardiac phase. Furthermore, imaging of the coronary arteries requires high spatial resolution and the reconstruction of thin (submillimeter) slices. Multiple cross-sections (typically ranging from 32 to 320, corresponding to 2.0 to 16-cm z-coverage) need to be acquired simultaneously so that the entire heart can be covered in approximately 10 seconds or less (well within one breath-hold). In response to these technical requirements, advances in CT hardware have been rapid and remain ongoing.

The Society of Cardiovascular Computed Tomography (SCCT) has formed a Basic and Emerging Sciences and Technology (BEST) Working Group to educate CT users on the fundamental technical topics of cardiovascular CT. The group is composed of physicians, technologists, and scientists working in the field of cardiovascular CT on scanners from multiple vendors. A review paper has been developed by this group to summarize the technical aspects of current state-of-the-art CT hardware and to describe scan modes on these systems for cardiovascular CT. The BEST Working Group conducted one in-person meeting and several telephone conferences and email exchanges to determine the content and focus of the document. Information is based on published literature and consensus about best practices. The final document was unanimously approved by the members of the BEST Working Group.

CT hardware

Table 1 summarizes state-of-the-art multidetector-row CT (MDCT) hardware for the main CT scanner manufacturers. Most scanners are described as single source and employ a single x-ray tube mounted on a gantry opposite a detector array. One scanner type, the dual-source CT scanner, has 2 x-ray tube/detector systems mounted on the same gantry, 95° apart.

Most scanners use ceramic (solid state) scintillation detectors coupled to photodiodes, which have improved spatial resolution and decreased noise compared to older xenon gas detector systems.^{1,2} The detectors are arranged in rows and columns; the number of active detector rows and the z-axis width of detectors in an array define the detector configuration. Detector configurations available on state-of-the-art scanners for cardiovascular imaging are illustrated in Figure 1.

The total nominal beam width (i.e., the total x-ray beam width per gantry rotation defined at the scanner isocenter) is determined by the detector configuration selected in a CT protocol. For example, a detector configuration of 64 × 0.5 mm corresponds to a total nominal beam width of 32 mm, and a detector configuration of 320 × 0.5 mm corresponds to a total

nominal beam width of 160 mm. The total nominal beam width for state-of-the-art scanners is listed in Table 1.

The detector row width defines the minimum thickness of the reconstructed CT image and is influenced by the z-axis dimension of the individual detector element.³ For example, a detector row width of 0.6 mm allows a slice thickness of 0.6 mm or greater. Through-plane or z-resolution is dictated by the z-axis width of individual detector elements. To achieve improved through-plane spatial resolution, some systems use an x-ray focal spot that alternates between 2 z-positions (ie, z-flying focal spot) to acquire 2 overlapping slices for each detector row.⁴ This feature has been reported to improve z-plane spatial resolution to 0.4 mm.⁵ In-plane spatial resolution is determined primarily by the number of x-ray projections available for reconstruction, the scan field of view, and the image matrix. The highest in-plane spatial resolution of current MDCT scanners has been reported to be in the range of 0.23 mm to 0.4 mm.^{5,6}

During helical scanning (described in detail below), some overscanning in the longitudinal direction (z-overscan) is required to ensure that sufficient data are available for reconstruction. Overscanning exposes organs adjacent to the desired scan range that are not of clinical interest. Dynamic or adaptive collimation is a hardware-based solution for collimating the x-ray beam such that extraneous radiation exposure is blocked by retractable collimator blades.^{7,8,9} Dynamic collimation has the greatest effect in reducing dose for shorter scan lengths and higher pitch values.

Z-overscanning can also occur during axial scanning. In this case, the amount of z-overscan depends on the planned scan length and the total beam collimation. If the planned scan length is not an integer multiple of the total beam collimation, the number of axial scans required to cover the anatomy will result in unnecessary exposure of organs adjacent to the range of clinical interest, as occurs with helical scanning. Adaptive collimation is a hardware solution offered on certain wide-detector array MDCT scanners; with this technique the detector collimation is automatically selected from a set of beam collimations in increments of 10 mm based on the planned scan length so as to minimize extraneous exposure.⁹ Adaptive collimation for axial scanning has the greatest effect in reducing dose with wide-detector array MDCT scanners because the portion of the total x-ray exposure in the axial scan mode attributed to z-overscanning increases with z-axis detector coverage.¹⁰

While early CT image reconstruction algorithms needed 360° of projection data to generate images, newer algorithms for cardiac imaging require only approximately 180° of data. For a single-source scanner, the time required to collect all data needed for reconstruction of cardiac images (ie, the acquisition time) is approximately one-half the gantry rotation time. The fastest single-source scanner currently available spins at 270 ms per rotation, for a nominal acquisition time and temporal resolution of approximately 135 ms. For a dual-source scanner, the acquisition time is approximately one-fourth the gantry rotation time, because the 2 x-ray source/detector systems collect data in half the time needed for a single x-ray source/detector array. The fastest dual-source scanner currently available spins at 280 ms per rotation, for a nominal acquisition time of approximately 70 ms.

In some instances, there is an opportunity to effectively improve temporal resolution during image reconstruction beyond the limits imposed by the gantry rotation time through the use of multicycle reconstruction. During helical scanning, attenuated photons passing through any given slice level of the patient's body strike each detector row in succession as the patient table moves through the gantry. Each consecutive detector row collects data from that particular slice of the patient's body, each at a slightly different time point within the cardiac cycle. However, if the heart rate is sufficiently high, the table speed sufficiently

slow, and the detector array sufficiently wide, the detector rows may collect data from the same location 2 or 3 times. Rather than reconstructing images from a single 180° arc of attenuation data obtained during 1 heartbeat, images can be reconstructed from 2 adjacent or overlapping arcs totaling 180° and obtained during 2 consecutive heartbeats.¹¹ For example, the first 90° of data might be obtained from the first heartbeat, and the second 90° might be obtained from the next heartbeat. Because the acquisition of data within each heartbeat is now occurring over a smaller scan angle, the time required for acquisition is shorter, effectively improving the temporal resolution. For cases in which exactly 90° of data is obtained from each cardiac cycle, for a single-source system, temporal resolution would effectively be improved 2-fold to a value equal to one-fourth of the gantry rotation time (eg, 67 ms for a gantry rotation time of 270 ms). Alternatively, 3 adjacent or overlapping datasets could be used from 3 consecutive cardiac cycles, again given a high enough heart rate and slow enough table speed, to effectively improve temporal resolution 3-fold up to a value equal to one-eighth of the gantry rotation time for a single-source system (eg, 45 ms for a gantry rotation time of 270 ms).

There are some caveats associated with multicycle reconstruction. Radiation exposure tends to be higher because of the requirement for overlapping x-ray exposure. In addition, the best temporal resolution can be achieved only at certain heart rates for which it is possible to acquire equal and spatially adjacent datasets from consecutive cardiac cycles (eg, scan angles equal to 0° to 90° from the first cardiac cycle and 90° to 180° from the second cardiac cycle). At most heart rates, datasets from consecutive cardiac cycles overlap, with 1 dataset spanning more than 90° and requiring a longer acquisition time. Finally, multicycle reconstruction requires a regular cardiac rhythm: the heart must come to rest in the same position, with the same cardiac cycle length, for every beat of the scan. Any variation, especially in cardiac cycle length, during the scan may result in motion artifacts. Still, the advantages of multicycle reconstruction are believed to outweigh the disadvantages for single-source systems and are automatically implemented for retrospective ECG-gated helical scanning on all CT systems.

Multicycle reconstruction is not limited to helical scanning. A scanner with a wide enough detector array to cover the entire heart in 1 rotation, such as a 320-row scanner, can image the entire heart in diastasis multiple times (up to 5 times) using prospective ECG-triggered axial techniques and can combine data to improve the effective temporal resolution. Adjacent data arcs can always be obtained, ensuring, for example, a 2-fold improvement in effective temporal resolution with 2-cycle reconstruction at all heart rates. However, a major disadvantage of using multicycle reconstruction with a wide detector array is the increase in radiation dose with repeated acquisitions; 2-cycle, 3-cycle, 4-cycle, or 5-cycle reconstruction results in a 2-, 3-, 4-, or 5-fold increase in radiation dose compared to single cycle reconstruction.

Scan modes

CT data are acquired using either a helical (also known as spiral) or axial scan mode. Helical data are acquired during continuous rotation of the gantry and simultaneous translation of the patient table. Axial data are typically acquired during a full (360°) or partial ($180^\circ + \text{fan-angle of the detector}$) rotation of the x-ray source and detector system around the patient while the patient table is stationary; if the z-axis coverage is insufficient to scan the entire region of interest, the patient table moves along the z-axis between periods of data acquisition.

For most cardiovascular CT indications, data acquisition or reconstruction is referenced to the cardiac cycle to restrict image data to a desired cardiac phase. This is accomplished by

using the patient's ECG signal to either prospectively trigger data acquisition or retrospectively gate data reconstruction. The resulting cardiac scan modes can be described as retrospective ECG-gated helical, prospective ECG-triggered axial, or prospective ECG-triggered helical modes.

Retrospective ECG-gated helical scan

With low-pitch helical scanning, data are retrospectively gated to the patient's recorded ECG signal. X-ray data are acquired throughout the cardiac cycle with continuous rotation of the gantry and movement of the table until the entire scan length is covered; the patient's ECG signal is simultaneously recorded. CT data are then retrospectively referenced to the recorded ECG signal, and images are reconstructed at desired time points within each cardiac cycle (Fig. 2). Retrospective ECG-gated helical scanning was the predominant scan mode for cardiac imaging with MDCT during its establishment as a standard clinical test.

Retrospectively ECG-gated techniques are less sensitive than prospectively ECG-triggered modes to arrhythmia; most scanner software allows for the deletion of data from premature ventricular beats, the insertion of undetected R-peaks, and the shifting of R-peaks to adjust for arrhythmia.¹² Therefore, retrospective ECG-gating may be preferred for patients with high and/or irregular heart rates.

The disadvantage of retrospective ECG-gated helical scanning is increased radiation dose from low-pitch application of x-rays during the entire cardiac cycle. However, because CT data are typically needed only from the cardiac phase with the least motion (eg, the mid-diastolic or end-systolic phase) for image reconstruction, a significant decrease in radiation dose can be achieved by modulating the tube current according to the patient's ECG signal to a maximum value during the desired reconstruction phase(s) of the cardiac cycle and a minimum value during the remaining phases (Fig. 2).¹³ Image quality is not compromised in appropriately selected patients (namely, coronary patients with stable sinus rhythm and most noncoronary cardiac patients).¹⁴ ECG-based tube current modulation, however, imposes limitations on helical imaging of patients with irregular heart rhythms for certain indications, primarily evaluation of the coronary arteries. Because ECG-based tube current modulation is prescribed based on an average of prescan R-R interval lengths, changes in heart rate during the scan could result in unintended lowering of the tube current during a desired phase of reconstruction for a given cardiac cycle. Additionally, because the x-ray tube cannot change its current instantaneously (i.e., there is an inherent delay in decreasing and increasing current), tube current modulation is less effective at higher heart rates.

Different strategies exist for ECG-based tube current modulation. Some CT scanners permit lengthening of the maximum tube current duration for patients with highly irregular heart rates.¹⁵ This expands the utility of ECG-based tube current modulation, but at the expense of increased radiation exposure. Some systems temporarily suspend or permanently switch off ECG-based tube current modulation if beat-to-beat variation exceeds a threshold value during data acquisition. The risk of improperly timed tube current modulation is then virtually eliminated in cases of severe arrhythmia, but again at the cost of increased radiation exposure.

The minimum tube current value applied during ECG-based modulation can range from zero¹⁶ to approximately 20%¹⁷ of the maximum value depending on the specific scanner type. Most CT manufacturers define a fixed value for the minimum tube current, but one manufacturer allows the user to choose a value equal to either 3% or 20% of the maximum value.¹⁸ Data are still available for reconstruction when a reduced (but nonzero) tube current is applied during a portion of the cardiac cycle, but image quality may not be sufficient for

the thin slices (eg, <1 mm) required for coronary evaluation. Reconstruction of thicker slices (e.g., 10 mm) for functional assessment may still be possible.¹⁹

Suggested heart rate thresholds for coronary CT angiography on select scanners are listed in Table 2. Because of the dose requirements for retrospective ECG-gated helical CT, the SCCT recommends reserving this mode for patients who do not qualify for prospective ECG-triggered scanning because of scanner-related limitations, irregular heart rhythm, and/or high heart rates.²⁰ Appropriate heart rates vary according to scanner model and clinical indication. Further, if retrospective ECG-gated helical CT is indicated, ECG-based tube current modulation should be used with the narrowest selectable window of maximum tube current except in patients with highly irregular heart rhythm.²⁰

Prospective ECG-triggered axial scan

Synchronization of axial scanning with the cardiac cycle is accomplished using the patient's ECG signal to prospectively trigger data acquisition. Scanning is initiated at a predefined time after the detection of an R peak (e.g., time when cardiac motion is minimal) while the patient table is stationary (Fig. 4).

For very wide detector array scanners (discussed in the next section), x-ray exposure during a single gantry rotation may be sufficient to cover the region-of-interest (e.g., for coronary artery imaging). For other scanners, the patient table must be moved to the next z-axis position while x-ray emission is stopped and data acquisition resumed once the table is stationary; the process of table movement/data acquisition is repeated until the entire scan length is covered (Fig. 4).

Prospective ECG-triggered axial scanning has emerged as a lower-dose alternative to retrospective ECG-gated helical scanning.²⁰ Axial imaging is restricted to scanners with at least 20 or preferably 40 mm of z-axis coverage per rotation because of long scan times as a result of moving the patient table between data acquisitions and the increased likelihood of misalignment artifacts.

For evaluation of the coronary arteries, a low and stable heart rate is typically necessary to minimize cardiac motion artifacts. Additional data beyond the minimum required for image reconstruction can be acquired with prospective ECG-triggered axial CT to permit retrospective adjustments of the reconstruction window, potentially reducing cardiac motion artifacts, but these adjustments come at the expense of increased radiation exposure.²¹ In an effort to prevent cardiac motion artifacts, many manufacturers offer automated arrhythmia rejection algorithms that postpone axial data acquisition until the heart rate stabilizes if an irregularity is detected (Fig. 3). Disadvantages of such algorithms include insufficient contrast enhancement in the case of long scan delays.

Specific heart rate threshold values for axial scanning vary according to scanner model and clinical indication; suggested values for coronary CT angiography on select scanners are listed in Table 2. Although heart rate requirements can be relaxed for noncoronary cardiac imaging because of less concern over motion artifacts, indications that require longer scan ranges (e.g., evaluation of the aorta) may not be appropriate for scanners with limited z-coverage per rotation (<40 mm).

Prospective ECG-triggered axial scanning typically does not allow for functional analysis because images can be reconstructed only during the prespecified phase of data acquisition. Some scanners do permit axial data acquisition during 2 phases (e.g., diastole and end systole) of the cardiac cycle, thus expanding the possibilities for some functional analysis but also increasing x-ray exposure.²²

SCCT guidelines recommend using prospective ECG-triggered axial techniques in patients who have regular, low heart rates.²⁰ Additionally, the width of the acquisition window for prospective ECG-triggered axial scanning should be kept as narrow as reasonably possible to acquire images of sufficient quality with the lowest radiation exposure.²⁰

Prospective ECG-triggered axial scan with wide detector arrays

Very wide detector array scanners allow prospective ECG-triggered axial data acquisition of the entire heart at a single time point with no table movement (Fig. 4). CT scanners with 320 detector rows can acquire a maximum of 16 cm along the z-axis per gantry rotation and z-axis coverage can be reduced with subsequent reduction in radiation dose.^{23,24,25} Acquisition of the entire heart at a single time point results in temporal uniformity of the image which offers the advantage of more uniform contrast enhancement and the absence of misalignment artifacts in the z-direction. Single heartbeat acquisition for coronary evaluation is typically applied to patients with regular, low heart rates (less than approximately 65–70 beats/min [bpm]) (Table 2). Patients with coronary artery disease who have higher heart rates are typically imaged on these systems by obtaining multiple prospective ECG-triggered axial data acquisitions at the same table position (covering the entire heart) during consecutive cardiac cycles and combining these data with multicycle reconstruction algorithms to improve effective temporal resolution.²⁶

Depending on the heart rate, the scanner automatically acquires data over 1, 2, or 3 cardiac cycles. However, this improved temporal resolution is achieved at the cost of significantly increased radiation exposure.^{23,24,27} Alternatively, the exposure window can be widened (eg, 40%–80% of the R-R cycle instead of 70%–80%) during axial imaging at the same table position, permitting image reconstruction across a wider range of cardiac phases. Radiation dose is also increased with this approach, but to a lesser degree than with multicycle acquisition.

If a longer scan range is required for patients with coronary bypass grafts or noncoronary cardiovascular indications, the 320-detector row scanner can be operated differently. The full detector array can be used in prospective ECG-triggered axial mode with table movement covering the anatomy of interest; typically 2 acquisitions are required to cover the entire thorax. Alternatively, a reduced number of detector rows can be employed (eg, 64, 80, 100, 128, or 160) and the scanner can be used in a standard prospective ECG-triggered axial mode, retrospective ECG-gated helical mode (with or without ECG-based tube current modulation), or retrospective ECG-gated medium-pitch (typically around 1) helical mode.

Prospective ECG-triggered helical scan

Dual-source scanners capable of achieving very high pitch values permit triggering of helical data acquisition by the patient's ECG signal.^{28,29} During prospective ECG-triggered high-pitch helical scanning, helical data acquisition is initiated by the patient's ECG signal and continues until the entire scan length is covered (Fig. 5). The acquisition time for a typical coronary CT angiogram is approximately 300 ms, allowing for data acquisition during a single cardiac phase in patients with low heart rates. As with prospective ECG-triggered axial techniques, a low (< 60 bpm) and stable heart rate is considered a prerequisite to minimize the risk of cardiac motion artifacts (Table 2). Additionally, scanning of the coronary arteries is limited to patients with lower body mass index (BMI) (< 30 kg/m²) because of field-of-view and x-ray output restrictions.³⁰ When prospective ECG-triggered high-pitch helical scanning is applied to imaging of the aorta, longer scan times spanning multiple cardiac cycles are required to accommodate longer scan lengths, but heart rate and size requirements are relaxed.^{31,32,33} For cardiovascular patients with higher heart rates or

higher BMIs, standard prospective ECG-triggered axial or retrospective ECG-gated helical scans should be performed (Table 2).

Another approach for prospective scan triggering in the helical mode is available with a 320-detector row scanner. With variable collimation, prospectively ECG-triggered scans can be acquired using helical acquisition with 64, 80, 100, and 160 slices per rotation. As with dual-source imaging, the helical pitch is faster than the helical pitch used in retrospectively ECG-gated scans, reducing scan time. Depending on the heart rate and exposure window, radiation doses can be reduced up to 80% compared to the doses used for retrospective ECG-gated helical data acquisition. The ability to select different z-axis coverage allows the examination to be tailored to clinical needs (e.g., smaller vs. large scan ranges). The advantage of helical over axial scanning with a 320-detector row system applies to scan ranges exceeding the z-axis coverage of a single rotation (ie, 16 cm): in this scenario, helical scanning avoids possible misregistration artifacts at the site where 2 axial datasets are merged. Particularly for patients with prior coronary artery bypass grafting, such an approach may be advantageous because the datasets frequently come together at points of interest (e.g., course of bypass grafts), which may cause ambiguity in interpretation.

Contrast Bolus Timing

There are two primary methods for timing scan acquisition during contrast enhancement of cardiac structures and vessels. The timing bolus method uses a small volume of contrast, typically 10 to 20 mL, injected during intermittent scanning of a region of interest to create a time-attenuation curve. The time of peak contrast enhancement is then used to determine the initiation of full cardiac scanning with respect to injection of a larger bolus of contrast. Alternatively, real-time bolus tracking allows for intermittent imaging of the region of interest during the injection of a large bolus of contrast. When the desired contrast enhancement is achieved, full cardiac imaging begins, usually after 1 to 8 seconds, depending on specific scanner settings.

Selection of the timing bolus or bolus tracking method is largely based on user preference, but certain scan modes benefit more from a particular method. For single heartbeat acquisition with a wide detector array scanner (eg, 320 detector row CT), bolus tracking is attractive because data from the entire heart can be acquired at a single time point within 1 to 2 seconds after the desired threshold of contrast enhancement is reached.

Dual-energy data acquisition

Dual-energy imaging describes the acquisition of 2 spectrally distinct attenuation datasets from the same region of interest. The high-energy dataset is acquired at 140 kV, and the low-energy dataset is acquired at either 80 or 100 kV, depending on the clinical question and patient size.

With current commercially available MDCT scanners, dual-energy data can be acquired using either 2 x-ray source/detector systems, each operated at different peak tube potentials,^{4,34,35} or using a single x-ray source/detector system with novel detector material that permits rapid kV switching.³⁶ A third approach for dual-energy imaging - that is not yet commercially available - uses a single x-ray source and dual layers of energy-sensitive detectors to acquire both low-energy and high-energy x-ray photons simultaneously.

Using 2 x-ray source/detector systems, dual-energy data are acquired with an approximately 95° separation between tubes, such that an approximately 75-ms time difference exists between acquisition of the 2 datasets at a given location. Using a single x-ray source/detector system with kV switching, dual-energy data are acquired with a 0.3- to 0.5-ms

separation.³⁷ With dual energy data acquisition, multiple monochromatic attenuation datasets can be generated and used to reconstruct multiple monochromatic image sets (ie, spectral images).^{38,39} For dual x-ray source and rapid kV switching, dual-energy data can be acquired in both axial and helical scan modes.

Several clinical applications of dual-energy CT acquisition modes have recently been reported.³⁷ Current cardiac applications include dual-energy myocardial perfusion and viability imaging and cardiac iron detection.^{40,41,42,43} Further potential cardiovascular applications include improved visualization of calcified coronary plaque and coronary stent patency.^{44,45} Dual-energy CT has also been used for calcified plaque removal in the carotid vessels.⁴⁶

Discussion

From the introduction of 2-slice CT scanners through the first generation of 64-slice scanners, the major manufacturers of CT scanners offered products with very similar technical specifications. Over the past 5 years, however, CT hardware design has diverged, with fundamental differences among scanners in terms of number of x-ray sources, detector geometry, and gantry rotation time. Contemporary cardiac CT scanners may incorporate one or 2 x-ray sources and detector arrays, an x-ray focal spot that is fixed or alternates between 2 z-positions, 32 to 320 detector rows, gantry rotation times from 270 to 350 ms, several different scan modes including some permitting single heart beat scanning, and a variety of approaches such as rapid tube potential switching with a single tube or dual tube potential application with two tubes that permit dual-energy imaging. Not covered in our overview, but complementing these hardware features, are numerous advances in CT software, including various image data-, projection data-, and model based- iterative image reconstruction techniques, and application-specific post-processing software.⁴⁷

The hardware features discussed have important implications in terms of the nature of image artifacts, radiation dose, and image quality in patients with high heart rates or irregular heart rhythms. These new features also allow for advanced CT imaging applications such as calcified plaque removal, myocardial perfusion, and viability assessment. No single scanner excels in every category; thus, the practitioner faced with selecting a new scanner needs to consider the technical features of each scanner, their impact on scanning in the patient population of interest, and cost. Physicians with the benefit of multiple cardiac-capable CT scanners should make the same considerations when triaging individual patients, to enable truly patient-centered cardiovascular CT imaging.

The significant advances in CT technology, and their diverse employment, offer the promise of future scanners incorporating more of these advanced features in a single platform along with newer technologies.^{48,49,50} Tomorrow's cardiac CT scanner will undoubtedly enable improved image quality under more robust conditions, more imaging applications, and lower radiation doses. Even so, as evidenced in this survey of CT hardware in early 2012, the field has clearly made remarkable technical progress from the days of the dynamic spatial reconstructor and EBCT, enabling the continued growth of cardiovascular CT and expanding its diagnostic and prognostic benefits.

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Abbreviations used

BEST	Basic and Emerging Sciences and Technology
CT	computed tomography
ECG	electrocardiogram
HU	Hounsfield unit
MDCT	multidetector-row computed tomography
SCCT	Society of Cardiovascular Computed Tomography

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Figure 1.
Schematic of the detector row configurations for current state-of-the-art multidetector row CT scanners from the major CT manufacturers.

Retrospective ECG-gated Helical

Figure 2.

Diagram showing retrospective ECG-gated helical CT data acquisition, with continuous gantry rotation and simultaneous patient table translation. Radiation dose from the scan is (A) highest with ECG-based tube current modulation turned off, (B) lower with the full tube current applied for only a portion (e.g., from 40%–80%) of the cardiac cycle, and (C) lowest with full tube current applied during only a single phase of the cardiac cycle.

Figure 3.

An example of an arrhythmia rejection algorithm. A schematic of ECG tracings from several cardiac cycles is shown. Based on the breath-hold exercise, the length of the first R-R interval was expected; however, the next R-R interval is much shorter as the result of a premature atrial complex. The system skipped the shorter cardiac cycle and initiated scanning in the next cycle with expected R-R interval.

Prospective ECG-triggered Axial

Figure 4.

Diagram showing prospective ECG-triggered axial CT data acquisition. Scan is initiated by the patient's ECG signal at a phase when cardiac motion is minimal while the patient table is stationary. Data are collected (**A**) over multiple heartbeats with most scanners, with the patient table advancing every other heartbeat and (**B**) during a single heartbeat with very wide detector array scanners [for HR 65 bpm (Table 2)].

Prospective ECG-gated Helical

Figure 5.

Diagram showing prospective ECG-triggered helical data acquisition. Helical data acquisition is initiated by the patient's ECG signal. Data are collected (A) over multiple heartbeats or (B) during a single heartbeat with dual-source scanners, which can achieve very high pitch (~ 3) values.

Table 1

Summary of CT hardware and parameters for state-of-the-art CT scanners from the major CT manufacturers

Scanner	X-ray sources, number	Detector rows, number	Detector-row x-axis dimension, mm [*]	Total nominal beam width, mm	Fastest gantry rotation time, s	Temporal resolution for each cross-sectional image, s [†]
GE Discovery CT750 HD	1	64	0.625	40	0.35	0.175
Hitachi SCENARIA	1	64	0.625	40	0.35	0.175
Phillips Brilliance iCT	1	128 [‡]	0.625	80	0.27	0.135
Siemens SOMATOM Definition FLASH	2 (95° apart)	64 [‡]	0.6	40	0.28	0.075
Toshiba Aquilion ONE	1	320	0.5	160	0.35	0.175

^{*} Values measured at scanner isocenter;

[†] Values do not reflect the use of multi-segment reconstruction;

[‡] Uses z-axis flying focal spot to sample each detector twice per rotation.

Table 2

Heart rate thresholds for scan mode selection for coronary angiography on state-of-the-art CT scanners

Scan mode	Heart rate threshold, bpm
Prospective ECG-triggered helical	60 bpm
Prospective ECG-triggered axial	65–80 bpm *
Retrospective ECG-gated helical	65–80 bpm *

* Threshold is higher for scanners with higher temporal resolution.

Bpm, beats per minute; CT, computed tomography; ECG, electrocardiogram.