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Cardiac CT: A system architecture study

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Abstract

BACKGROUND: We are interested in exploring dedicated, high-performance cardiac CT systems optimized to provide the best tradeoff between system cost, image quality, and radiation dose.

OBJECTIVE: We sought to identify and evaluate a broad range of CT architectures that could provide an optimal, dedicated cardiac CT solution.

METHODS: We identified and evaluated thirty candidate architectures using consistent design choices. We defined specific evaluation metrics related to cost and performance. We then scored the candidates versus the defined metrics. Lastly, we applied a weighting system to combine scores for all metrics into a single overall score for each architecture. CT experts with backgrounds in cardiovascular radiology, x-ray physics, CT hardware and CT algorithms performed the scoring and weighting.

RESULTS: We found nearly a twofold difference between the most and the least promising candidate architectures. Architectures employed by contemporary commercial diagnostic CT systems were among the highest-scoring candidates. We identified six architectures that show sufficient promise to merit further in-depth analysis and comparison.

CONCLUSION: Our results suggest that contemporary diagnostic CT system architectures outperform most other candidates that we evaluated, but the results for a few alternatives were relatively close. We selected six representative high-scoring candidates for more detailed design and further comparative evaluation.

Keywords

Computed tomography; CT; cardiovascular disease; cardiovascular imaging

1. Introduction

According to the World Health Organization, cardiovascular disease (CVD) is the leading cause of death globally, with an estimated 17.3 million CVD-related deaths in 2008 [1]. The American Heart Association estimates that, in the US alone, there were 386,324 coronary heart disease (CHD) related deaths in 2009; each year 635,000 Americans have a new heart attack, an additional 150,000 have a silent first myocardial infarction, and 280,000 have a recurrent attack [2]. The total number of inpatient cardiovascular operations and procedures increased from 5.9 million in 2000 to 7.6 million in 2010, and the AHA estimates total 2009 direct and indirect cost of CHD and stroke to be \$312.6 billion in the US. In Europe, the annual incidence of acute coronary syndrome (ACS) is estimated at between 1 per 80 to 1 per 170 of the population per year, while the incidence of chest pain leading to in-patient admission for suspected ACS is much higher [3]. The Global Register of Acute Coronary Events (GRACE) reviewed data from 11,543 patients who were admitted with the suspicion of ACS, obtained from 18 population-based clusters in 14 countries worldwide, and showed that unstable angina was the most frequent clinical presentation at admission and confirmed at discharge [4].

In the early 1990's, it became clear that a majority of myocardial infarctions did not originate at stenotic locations. Clinical trials of thrombolytic agents in acute myocardial infarction revealed that the lesion underlying the clot was often not stenotic [5]. More recent results from the "Courage" trial comparing therapy for stable coronary artery diseases to percutaneous coronary intervention indicated that "fixing the stenosis" did not reduce death and other major cardiac events [6]. The current research emphasis is thus to improve our understanding of the pathobiology and genetics of coronary artery diseases. Given the global prevalence, high mortality and impacts of the diseases, better detection of atherosclerosis before symptom presentation is imperative.

Catheter coronary angiography (CCA), an invasive x-ray imaging modality, has been the clinical standard for identifying flow-limiting lesions in coronary arteries. Recently, other invasive imaging techniques such as intra-coronary ultrasound (ICUS) and optical coherence tomography (OCT) have entered this arena to supplement CCA. As alternatives, computed tomography (CT) and magnetic resonance (MR) have been used to *non-invasively* diagnose the presence of atherosclerosis even in the absence of stenoses. Widespread availability of the multi-row/cone-beam CT scanner, its non-invasive nature and short scan time have led to the emergence of coronary CT angiography (CCTA) as a most promising tool for noninvasive coronary angiography [7–9]. CCTA has made imaging of the lumen of the coronary arteries possible with close correlation to CCA and excellent diagnostic accuracy [10]. Furthermore, preliminary results have also demonstrated that CT angiograms have good correlation with ICUS in estimating plaque volume [11] and can classify the plaques into lipid-containing, fibrotic and calcified but with significant overlaps [12].

The two major advances that were key to the recent growth in CCTA are faster rotation speed and multi-slice/cone-beam geometry [13–16], essentially eliminating the tradeoff between slice thickness and z-coverage. A dual-source CT scanner [17] and a 320-slice CT scanner [18] were introduced to achieve temporal resolution of about 100 ms and coverage of 16 cm respectively. However, an ideal cardiac CT scanner would combine all performance characteristics.

Despite the impressive advancement in CT technology, there remains a critical need for cardiac CT with significantly better image quality at lower radiation dose. Therefore, we embarked on a project to identify the key clinical requirements for a cardiac-specific CT scanner (summarized in Section 2.1), identify a wide range of possible CT system architectures to meet the clinical requirements considering all relevant contemporary and emerging technologies (summarized in Section 2.2), develop a methodology to comparatively evaluate these architectural alternatives (summarized in Section 2.3), and select a few of the most promising candidate architectures for further analysis and development (summarized in Section 3).

2. Methods

2.1. Performance requirements

To begin our study, we revisited the clinical applications of cardiac CT and identified the key clinical performance requirements, based on literature review and discussions with clinical experts.

The two most common tasks for cardiac CT are coronary artery calcium scoring (CACS) and coronary artery imaging (CAI). A non-contrast-enhanced CACS scan can produce a calcium score as an estimate of the calcified plaque in the coronary arteries. The ACCF/AHA 2007 Clinical Expert Consensus Document concluded that it should be reasonable to consider the use of CACS in asymptomatic patients with intermediate CHD [19]. CAI is a contrast-enhanced exam that assesses the vessel structures, and more particularly identifies the extent of vessel narrowing (stenosis). Accurate assessment of the calcium burden and vascular stenosis requires sufficient spatial, contrast, and temporal resolution. Enabling accurate and robust identification of positive remodeling and plaque consistency poses even more challenging requirements for cardiac CT. Moreover, applications that require full-organ coverage in a single rotation, such as myocardial perfusion studies, necessitate extended volumetric coverage.

We identified specific performance objectives, as summarized in Table 1. Modern CT scanners have gantry rotation times on the order of 300 ms. Electron-beam CT (EBCT) was the first dedicated cardiac CT imaging system with temporal resolution as low as 50 ms, but it has essentially become obsolete because of its high cost and major limitations in imaging performance. We reached a consensus that 50 ms is a good and realistic target for temporal resolution [20], whether achieved through hardware, smart algorithms or a combination of both. Note that the actual the temporal resolution varies across the FOV; the temporal resolution cited is in the center of the FOV and represents the nominal. In this article, we consistently cite the center, i.e. nominal temporal resolution. Volumetric coverage in modern

cardiac CT scanners ranges from 40 mm to 160 mm. We set 160 mm as our target since this is sufficient to cover the entire heart in one rotation for the majority of the patient population. Transaxially, the heart anatomy only requires a 200 mm diameter, assuming it is perfectly centered at the scanner rotation axis. To avoid complications with cardiac centering, we required a field-of-view diameter of 300 mm. The other performance requirements are highly dependent on the particular cardiac CT exam. We require an isotropic spatial resolution of 20 lp/cm (5% cutoff of the modulation transfer function) for CAI, no more than 10 HU noise standard deviation for cardiac perfusion, no measurable image artifacts and less than 1 mSv effective dose, consistent with the recent sub-mSv CT vision of the National Institutes of Health. Some of the above clinical requirements can be directly translated to system design specifications for the candidate architectures. Other requirements will have to be evaluated through a detailed simulation and reconstruction study, which is beyond the scope of this paper.

All input design parameters are summarized in Table 2. In most cases we assume a rotation time of 0.2 s, which our mechanical experts believe is a challenging but achievable limit. For the one architecture where only the detector is rotating, a rotation time of 0.15 s is assumed. Geometrically, we compare all architectures at the same source-to-isocenter distance (540 mm) and source-to-detector distance (950 mm). A spatial resolution of 20 lp/cm is slightly higher than today's state-of-the-art systems with 0.6 mm detector cell pitch (at isocenter), hence we uniformly assume a 0.5 mm detector cell pitch (at isocenter) and a focal spot size that is consistent with this. Whether these numbers result in precisely 20 lp/cm isn't critical for the purpose of this work, since this is a comparative study and all candidate architecture will produce the same spatial resolution given our uniform design choices. Similarly, dose efficiency is only evaluated qualitatively in this study, comparing the relative pros and cons of different architectures, but not evaluating whether the absolute effective dose number is achieved. Finally, for this comparative study, we assume 100 kW power for each rotating-anode x-ray tube and 50 kW power for stationary distributed x-ray sources.

In summary, since the goal of this paper is to perform a fair, high-level comparison between a large number of architectures, we uniformly adopt a number of reasonable design choices. However, the precise design of the CT systems and detailed evaluation of their resolution, noise and dose characteristics is left for a future study, which we will perform based on a limited number of high-scoring architectures selected in this study.

2.2. CT system architectures

We performed a broad search for candidate architectures by reviewing literature and through in-depth workouts with experts in this field. We included contributors from academia and industrial research. We applied Triz methods [21] in order to foster generation of new ideas from an objective, requirements-driven perspective. During this process we developed a functional requirements diagram shown in Fig. 1.

Each candidate architecture was assigned a reference acronym, as follows:

$\langle \text{system type} \rangle - \langle \text{size} \rangle \text{FOV}$, where $\langle \text{system type} \rangle$ is a unique acronym describing the group of architectures, and $\langle \text{size} \rangle$ refers to the transaxial field of view (FOV) and is either F for

full (500 mm), C for cardiac (300 mm), or M for mixed (for example, a dual-source system where one imaging chain provides a full FOV and another imaging chain provides a cardiac FOV).

The candidate architectures are summarized in Table 3. We produced simple sketches of each proposed architecture, to help us easily visualize the significant aspects of each.

The most basic architecture that we identified is a standard single-source CT (SSCT) system providing a full FOV. We defined this as the “baseline” with which we would compare other candidates. The baseline is designated SSCT-FFOV.

Many architectures differed only in the size of the transaxial FOV, or in the combination of sizes (for multiple beam-line systems, e.g. one FFOV and two CFOV beam-lines). As summarized in Table 2, certain constants were incorporated in all architectures. As explained in the previous section, at this stage, no attempt was made to perform a detailed design such that the systems would match the specific spatial resolution, noise and dose targets, but we focused on analyzing how favorably each architecture trades off all the requirements.

For each architecture, we considered various modes of operation that could be applied to cardiac scanning (full scan, half scan, various timing options for multiple-source systems, etc.) and chose one which we felt would be the “preferred operational mode” for cardiac scans. We performed most evaluations while considering each system operating in this preferred mode. We also assumed that each system was collimated to operate in CFOV mode – regardless of FFOV capability – to avoid unnecessarily increasing the radiation dose.

Many of these architectures employ a fixed CFOV. For cardiac scans on FFOV and MFOV architectures, we envision adjustable transaxial collimation of the x-ray source, to reduce the transaxial FOV from FFOV to CFOV, and thereby reduce radiation dose and x-ray scatter. For all architectures operating in CFOV mode, projection data would be truncated transaxially (no data is measured outside the CFOV). For this to be feasible, an “interior” image reconstruction algorithm is required. Such an algorithm needs to produce artifact-free reconstructions from truncated projections. Recent developments [22–25] suggest that this is realistic; we have assumed the existence of such an algorithm, and all evaluation was performed assuming the system was operated in a CFOV mode.

We now provide a description of the different groups of architectures and some representative architectures, referring to Table 3 for a complete list of architectures.

2.2.1. Conventional source, conventional detector architectures—All architectures in this group have focal spots at only one longitudinal position, which leads to a large 17° cone angle and commensurately high potential for cone-beam artifacts. Therefore, for any system in this group to succeed, advanced wide-cone image reconstruction algorithms would be required.

SSCT-FFOV (Fig. 2a), our “baseline” architecture, is the logical extension of a typical contemporary commercial diagnostic CT system, with the detector size increased to meet the longitudinal coverage goal, and the rotation period reduced to 200 ms, driven by the

temporal resolution goal. A single-source system with this rotation period has a nominal temporal resolution of 100 ms, and will only meet the 50 ms temporal resolution goal if sophisticated motion correction algorithms are applied. Recent developments demonstrate that this is realistic [26–28].

This architecture benefits from a full FOV, which permits use as a general-purpose scanner. In addition, CFOV projection data can be acquired by applying optional transaxial collimation, thus truncating the projections laterally, and reducing the radiation dose to the patient. For most analysis, we considered this system operating in CFOV half-scan mode, i.e. rotating for $180^\circ + \text{the } 32^\circ \text{ fan angle}$. In this preferred mode, interior reconstruction is required. However, it would not be required if the transaxial collimation is not applied, and the system is run in FFOV mode (projection data acquired while rotating for $180^\circ + \text{the } 55^\circ \text{ fan angle}$).

We also considered a cardiac-specific version, **SSCT-CFOV** (Fig. 2b). This would have a lower production cost, but would be limited to applications such as cardiac and head scanning, where a CFOV is sufficient.

DSCT-MFOV (Fig. 3a), dual-source CT with mixed field-of-views, has two beam lines: One with a FFOV detector and one with a CFOV detector. This is related to the commercial Siemens Definition Flash system [29], but our proposed architecture has 160 mm longitudinal coverage and 0.2 s rotation period. Like SSCT-FFOV, it can be used as a general-purpose FFOV scanner, for cardiac scanning with the FFOV, or it can be operated with transaxial collimation of the FFOV beam line to the CFOV. Nominal temporal resolution can be approximately 50 ms, however this would be slightly degraded due to the $>90^\circ$ offset between beam lines, necessitated to avoid interference between the CFOV detector and the FFOV x-ray fan beam. For the sake of this high-level comparison we ignore this additional 7° offset in the following analyses. For most analysis, we considered this system operating in a preferred mode using the CFOV in half-scan mode. Projection data would be acquired over a range of $90^\circ + \text{the } 32^\circ \text{ fan angle}$. In this mode, interior reconstruction is required. However, if the transaxial collimation is not applied, and this system was operated MFOV (one beam line in FFOV and the other beam line in CFOV) in half-scan mode, the FFOV detector would provide some information outside the CFOV, and interior reconstruction would not be required. However, the FFOV beam line would have to keep scanning for $180^\circ + \text{the } 32^\circ \text{ fan angle}$ and the FFOV image data would not be available at high temporal resolution. Finally, this system can be operated in FFOV mode by using only the FFOV detector. This mode is identical to SSCT-FFOV operation in FFOV mode.

We also considered a completely FFOV variant (**DSCT-FFOV**, not shown) and a cardiac-specific version, **DSCT-CFOV** (Fig. 3b). DSCT-CFOV would have a slightly lower production cost, but would be limited to applications such as cardiac and head scanning, where a reduced FOV is sufficient. With the current geometry, DSCT-FFOV is impractical since there is insufficient room for two full beam lines (the two detectors would overlap); this could be resolved by adjusting the system geometry, but then the system size would become impractical. Due to the dual beam lines, these systems all closely approach the

temporal resolution goal of 50 ms, with no special motion-compensating algorithms required.

First proposed in 1980 [30], triple-source CT (TSCT) is a more ideal multiple-source option than DSCT, but obviously at higher cost. **TSCT-CFOV** (Fig. 4a) is one variant, which can acquire complete data by rotating for $60^\circ +$ the 32° fan angle, and would provide a nominal 33 ms temporal resolution, but would require interior reconstruction. Alternatively, it can acquire data over a 120° rotation interval, and provide “full scan” projection data. We hypothesize that intelligent algorithms could preserve the inherent 33 ms temporal resolution of the half-scan data while combining the noise benefit of the full-scan data, for example by estimating the motion and averaging the motion corrected images; this is a topic for future research. Other variants include **TSCT-FFOV**, a version with three FFOV beam lines (not shown); **TSCT-MFOV**, a version with one FFOV beam line (Fig. 4b); and **TSCT-CFOV(A)**, similar to TSCT-CFOV, but with one beam line “flipped” – source and detector interchanged, making it asymmetric (A) (Fig. 4c). With the current geometry, TSCT-FFOV and TSCT-CFOV(A) are impractical due to insufficient room in the gantry – these systems couldn’t be implemented with the geometry specified, since the components would overlap, as illustrated in Fig. 4c. This could be resolved by adjusting the system geometry, for example by increasing the source-to-isocenter distance and reducing the detector size, but the system size would make it more complex, and the 0.2 s rotation time would become even more challenging. TSCT-MFOV would provide an interesting tradeoff: A system with CFOV capability and 33 ms temporal resolution, as well as FFOV capability for general-purpose scanning. However, this configuration can’t be implemented exactly as shown – the two sources near the FFOV detector are too close to the detector for practical implementation and would need to be moved away from the detector. That would result in a non-uniform angular arrangement, which would compromise the attractive aspect of operational mode flexibility as previously mentioned regarding TSCT-CFOV.

Finally, we considered **QSCT-CFOV** (Fig. 4d), quintuple source CT. With the current geometry shown, this is impractical due to insufficient room in the gantry. This could be resolved by adjusting the system geometry but the same issues would arise as were discussed regarding TSCT-FFOV and TSCT-CFOV(A), and this architecture with 5 x-ray tubes and 5 detectors would have extremely high cost and complexity.

2.2.2. “Twin Z” source, conventional detector architectures—All architectures in this group have sources with a longitudinally offset pair of focal spots (i.e.: Stacked in z), which leads to smaller 11° or 12° cone angles and reduced potential for cone-beam artifacts. The two x-ray sources can either be incorporated in one vacuum housing, as dual anodes on a common shaft, or as two independent x-ray tubes with the anodes back to back. In either case, the x-ray sources would be operated in an alternating manner (only one activated at any time). The longitudinal offset between the two focal spots of 128 mm was chosen such that the two cones intersect outside the CFOV, while minimizing the detector size. The two cone beams don’t completely overlap (Fig. 5a, right and Fig. 5b, right), leading to lower fluence in some regions of the volume of interest (VOI), and consequently lower image quality (higher x-ray noise).

TZCT-FFOV, or twin-z CT, is shown in Fig. 5a. Like SSCT, nominal temporal resolution would be approximately 100 ms, and motion correction would be required to achieve our target temporal resolution.

DTZCT-CFOV, or dual twin-z CT, is shown in Fig. 5b. Like DSCT, nominal temporal resolution can be approximately 50 ms, with projection data acquired over a rotation of 90° + the 32° fan angle. In this mode, interior reconstruction is required: There is no projection data outside the CFOV.

We also considered other variants: **TZCT-CFOV**, and **DTZCT-FFOV**, and **DTZCT-MFOV** (none shown).

2.2.3. Linear source, conventional detector architectures—All architectures in this group are an extension of the Twin Z concept, having three (or more) longitudinally offset focal spots for each transaxial source position, leading to further reduced potential for cone-beam artifacts. Most likely, the x-ray sources would need to be incorporated in one housing, as multiple anodes on a common shaft [31, 32], or as stationary anodes [33]. Typically, the x-ray sources would be operated in a sequential manner (only one spot activated at any time), leading to less than full flux in some regions of the VOI, and consequently lower image quality (higher x-ray noise). This effect is exacerbated compared to Twin Z architectures. The longitudinal offset (80 mm for a total of three longitudinally offset focal spots) was chosen such that the outer focal spots aligned with the edge of the longitudinal coverage (160 mm).

We considered single-source and dual-source variants, incorporating various FOVs: **LSCT-FFOV** (Fig. 5c), **LSCT-CFOV** (not shown), **DLSCT-MFOV** (not shown), and **DLSCT-CFOV** (Fig. 5d). We decided not to exhaustively include combinations such as triple and quintuple source versions at this stage of the investigation. In case linear sources proved promising, we could broaden the design space again in future work.

2.2.4. 2D source (Inverse geometry) architectures—All architectures in this group have an x-ray source that contains a 2D array of focal spots [34–36]. These focal spots are scanned during one view period (typically only one spot activated at any time) in a raster or arbitrary sequence. This geometry permits the use of a much smaller detector, thus trading off the cost of the detector for that of a more complex source. Potential benefits include reduced cone-beam effects, and the ability to modulate the flux in the transaxial direction, resulting in an “electronic bowtie” that can be customized to the actual patient, for improved dose reduction. The required multi-source technology has been successfully demonstrated by our group in a prototype gantry-based CT system [37, 38].

IGCT-CFOV (Fig. 6a) provides a nominal 100 ms temporal resolution. This architecture would require both intelligent motion correction algorithms and interior reconstruction algorithms to meet cardiac temporal resolution requirements. Also, multi-source technology must be further developed to provide sufficient flux for low quantum noise at high speed. Another drawback to this architecture is that the X rays expose the detector from many different angles, which means that an anti-scatter grid cannot be used, increasing its

susceptibility to scatter. The mean scatter contribution can be corrected, but the noise associated with the scatter cannot be removed, resulting in higher noise in reconstructed images. The increased susceptibility is partly compensated by the smaller fan angle of each transaxial focal spot causing far fewer scattered x-rays to arrive at the detector, as well as by an increased detector fill factor resulting in a higher detection efficiency. In comparison, all previously discussed designs can use a focally aligned anti-scatter grid at the detector, at least in the transaxial direction, to remove scatter introduced by the subject.

We also considered a FFOV variant, **IGCT-FFOV** (Fig. 6b).

2.2.5. Fully stationary CT architectures—Fixed-anode, multi-spot source technology leads to the possibility of a fully stationary CT (SCT) system, which can provide extremely high temporal resolution. A larger number of variations on this concept have been proposed by our team [39]. These architectures could provide coverage of the entire heart, with temporal resolution of 50 ms or better.

For these architectures to be viable, a geometry must be envisioned that provides sufficiently complete projection data throughout the VOI. Of the many systems that we have considered, all have fundamental limitations regarding data completeness.

SCT with a single source ring centered longitudinally on the detector (**SCT-SPHSR-CFOV**, Fig. 7a, where SPHSR stands for single peek hole source ring) must have a “peek hole” cut through the center of the detector to provide VOI illumination from the x-ray source on the opposite side of the detector. Therefore, there is no data available in the central planes of the VOI. In order to avoid this, one can envision a single longitudinally offset source ring (**SCT-SOSA-CFOV**, Fig. 7b, where SOSA stands for single offset source arc), similar to the Imatron design [40], but with 160 mm longitudinal coverage. This topology results in a large asymmetrical cone beam, which generates asymmetrically incomplete data in the longitudinal direction. We can attempt to produce more symmetrically complete data by employing two offset source rings (**SCT-DOSR-CFOV**, Fig. 7c, where DOSR stands for dual offset source rings). However, no projection data is available in the center of the VOI, leaving a “hole” in that critical region of the reconstructed image volume.

A further drawback of these architectures is the high cost of the large source and of the large detector. More variations with partial or full source and detector rings are conceivable but are not exhaustively included in this high-level analysis.

2.2.6. Semi-stationary architectures—We can conceive of a multi-spot stationary source surrounding a rotating detector. Due to the lower rotating mass, this system could be operated at a somewhat higher rotation speed than an equivalent rotating source system. Such architectures could perhaps provide an interesting compromise – a system with high temporal resolution, but without the high detector cost of a fully stationary system.

We estimated that the gantry rotation time could be 150 ms. Therefore, temporal resolution of a single-detector system operating in half-scan mode could be 75 ms, and a dual-detector system could achieve 37.5 ms temporal resolution. The dual-detector version, ring-source

with dual rotating detector or **OSDRDCT-CFOV**, is shown in Fig. 8a. A version that employs an inverse-geometry detector, **OSRDIGCT-CFOV**, is shown in Fig. 8b.

For completeness, we also considered the opposite arrangement, using a rotating source (RS) and a stationary ring detector (OD). **RSODCT-CFOV** (Fig. 8c) includes the worst aspects of all components: a very large and expensive detector, and a rotating gantry that is speed-limited by a rotating-anode source.

2.2.7. Novel architectures—Two architectures that we identified were quite unique.

Sector Source CT, designated **CSCT-CFOV** (Fig. 9a) uses a completely different approach to achieving high temporal resolution. The first C in CSCT refers to the arc- or C-shape of the x-ray source. By rotating the gantry very slowly (on the order of 10 s for one full rotation) and acquiring projections from a range of angles almost instantaneously (activating different focal spots in rapid succession), using a multi-spot source, this architecture would, in principle, provide sufficient data to perform 4D reconstruction with temporal resolution in the range of 1 to 10 ms, or less. The basic idea is that the gantry would rotate over less than the length of the source arc during one heartbeat, such that we can synthesize a complete dataset from very fast partial scans obtained over multiple heartbeats. The primary drawback to this concept is that it assumes positional repeatability of heart anatomy over multiple cardiac cycles (spatial position of the heart relative to temporal position in cardiac cycle). This *repeatability* does not necessarily mean heart motion *periodicity*. However, we believe that this assumption is probably unrealistic, especially for patients with heart disease.

The final concept that we considered is Concentric Circling CT, **CCCT-CFOV** (Fig. 9b). This involves rotating both the source and detector around an axis perpendicular to the axis of gantry rotation, while simultaneously revolving the gantry around the patient. Although scientifically interesting, it is mechanically impractical to consider producing the required motion trajectory for these heavy components, at the required velocity.

While we covered a wide range of CT architectures, this study is not exhaustive. For instance, concepts were proposed with alternate detector designs [41, 42] or with source and detector rotating in opposite directions [43, 44].

2.2.8. Other system features—

We also considered a number of features that could be applied to some or all of the above architectures. We didn't include these in our comparative analysis; however, in future work we will consider applying these in order to produce a more optimal cardiac-specific scanner. These include: Spectral CT (multi-kVp source or spectrally sensitive detector); flat-panel detectors, hybrid detectors (e.g. a detector with varying pixel size) and aggressive bowtie filters (strongly attenuating all x-rays outside the CFOV).

2.3. Evaluation methods and metrics

All candidate architectures were evaluated and compared in terms of temporal resolution, dose-efficiency, data completeness, power capability, cost and complexity. The following sections describe each metric in more detail.

Our goal was to keep this screening architecture study very objective, while it is necessarily based on a lot of expert judgment. Some aspects of the proposed scoring and selection process are based on straightforward calculations. Some other aspects simply have to rely on judgment by CT experts. Hence, it is understood that the results will not perfectly match the opinion of any single person but represent the consensus between a large number of CT experts with a variety of backgrounds.

2.3.1. Cost model—We developed a cost model, estimated the cost of manufacturing each architecture, and normalized it to the baseline system. Estimated costs were based on known per-unit manufacturing costs for existing technologies and estimated per-unit costs for emerging technologies (units being number and size of x-ray source housings, number of focal spots, number of detector pixels, area of detector, mass of rotating components, etc.)

The absolute cost of components is proprietary information that is closely guarded by CT system manufacturers. However, it is widely understood that the detector is the dominant cost in contemporary CT systems and represents roughly half of the total cost of the system; whereas the X-ray source might represent approximately 10% of the total system cost. This generalized cost trend is captured in the relative cost estimates shown in Table 3, including a higher incremental system cost for system concepts utilizing full field-of-view detectors or multiple detectors. Therefore, if detectors increase in size in order to provide higher longitudinal coverage, or multiple detectors are used, the system cost becomes even more heavily dominated by the detector cost. Considering this trend, one can appreciate the desire to identify an architecture that uses a smaller detector, and still satisfies all performance requirements.

In order to combine all comparative results into a net score for each architecture, we needed all metrics to be rated on positive criteria, so we converted the cost score to a “value” score, equal to the scaled difference between the system’s cost and the maximum system cost.

2.3.2. Estimated system performance, simplicity, and capability—In this work we performed a high-level qualitative analysis. However, we felt that it was important to attempt to quantify the performance, simplicity and capability tradeoffs between the candidate architectures, in hopes that this process would illuminate a candidate system to be stronger or weaker than our intuitive understanding led us to believe. Therefore, we identified a few critical parameters and we assigned system design experts to each parameter. These experts proposed specific criteria that would affect each parameter, and for each criterion, award/penalty points associated with specific system characteristics. The award/penalty points were intended to linearly relate each system’s performance for that parameter, without considering the zero value for that scale. As a group, we reviewed and refined the proposed scoring system. The assigned system design experts then rated each candidate architecture using the defined point system, and the results were reviewed and discussed by the group.

Table 4 shows the selected parameters, rating criteria, and award/penalty points for each criterion. Note that award/penalty points shown in Table 4 were determined for each specific

candidate architecture defined in Table 3. In all cases, the scoring scheme was adjusted to keep all raw scores in the range of 0–10.

Since we set spatial resolution as a constant across all systems, we reduced our evaluation of “Performance” to parameters related to image quality (IQ), temporal resolution, and radiation dose.

- “IQ - artifacts - raw data completeness” refers to the completeness of data in the CFOV – “interior” reconstruction is assumed for all architectures – and the scoring system is relative to completeness of the baseline system in CFOV mode. Data completeness can be precisely estimated for a given voxel using the method proposed by Liu [45]. For reference, the baseline system has a Radon completeness of 100% in the mid-plane and 99.9% at the largest cone-angles and receives a score of 5, as do all other rotating systems without longitudinally offset sources. Systems with a reduced maximum cone angle (TZCT, LSCT, IGCT, CCCT) receive a score of 6. Systems with an increased maximum cone angle (SCT-SOSA) receive a score of 4. Systems with a small region that is not sampled (completeness of 0%) (SCT-DOSR) receive a score of 3. Systems with an entire slab missing (SCT-SPHSR) receive a score of 1.
- “IQ - noise – x-ray fluence capability” refers to the architecture’s ability to produce x-ray fluence sufficient to produce images with noise at the same level as the baseline system. In order to evaluate architecture’s ability to achieve the absolute noise target given in Table 1, a detailed analysis would be required, including analysis of yet-to-be developed reconstruction algorithms. This is beyond the scope of this work. Therefore, we chose to use x-ray fluence (x-ray source power combined with scan time and multi-source operation) as a surrogate method to evaluate architecture’s ability to achieve the noise performance target. Scan time for a cardiac scan depends primarily on rotation time, number of beamlines, and the rotation interval used by the reconstruction. For instance, an SCT system has a tube power of 50 kW and can acquire data for 50 ms, resulting in 2.5 kW·s. Systems with an effective integrated power of 10 kW·s (SSCT, DSCT) receive a score of 7. Systems with effective integrated power of 20 kW·s or more (TSCT, QSCT) receive a score of 9, 7.5 kW·s (OSDRD) a score of 6, 5 kW·s (TZCT, LSCT) a score of 5, 3.75 kW·s (OSRD) a score of 4, 2.5 kW·s (SCT-*, CCCT) a score of 3 and < 2.5 kW·s (*IGCT) a score of 2.
- “Temporal resolution capability” refers to the architecture’s intrinsic ability to provide temporal resolution benefit relative to the baseline system. Temporal resolution is inherently non-uniform across the transaxial FOV, so we chose to estimate this at the axis of rotation. The numbers do not reflect software motion compensation methods. It’s an open question whether those methods can match the performance of the inherently more costly hardware approaches like those evaluated in this work. Therefore we gave an advantage to systems that can achieve 50 ms in hardware compared to systems that can achieve 100 ms in hardware. Systems with a temporal resolution of 100 ms receive a score of 1, 75

ms a score of 1.5, 50 ms a score of 2, 33 ms a score of 3, 25 ms a score of 4 and systems that combine multiple heart beats a score of 0.

- “Dose efficiency” refers to three considerations: 1) the architecture’s delivery of radiation outside the VOI, which results in penalty points, 2) the uniform (award points) or non-uniform (penalty points) delivery of radiation within the VOI, and 3) the anticipated x-ray scatter-to-primary ratio (SPR), since increased noise from scatter would result in an increase in dose, assuming constant image noise. All single beamline systems receive a score of 7. The additional cross-scatter between beamlines and the necessary overlap to stitch partial sinogram together justify a score of 6 for two beamlines, 5.5 for three beamlines and 5 for five beamlines. Systems with most severe scatter challenges receive a dose-efficiency score of 2.

The “Simplicity” category includes parameters that are related to cost but are non-recurring expenses, which would be incurred in order to develop and commercialize the technology required to design the system and put the design into production. The parameters we evaluated are System simplicity, Calibration simplicity, and Reconstruction simplicity. System simplicity scores were derived from discussions with CT hardware integration experts and reflect the relative complexities associated with novel x-ray source designs, limited space in the gantry, and system weight. Similarly, calibration simplicity scores were derived from discussions with CT calibration experts and reflect the relative complexities associated with eliminating scatter artifacts, spectral and x-ray intensity calibration, and geometric calibration. For reconstruction simplicity, systems for which no known reconstruction algorithms can directly be applied receive 1 point penalty.

The “FFOV capability” parameter rewards some architectures for having the flexibility to serve as a general-purpose scanner, and perform FFOV scanning. However, all performance analysis was considered when operating the scanner in CFOV mode, regardless of FFOV capability.

2.3.3. Parameter normalization/weighting—Because our selected parameters were scored on an arbitrary scale, we needed to determine a reasonable minimum (lower limit) and maximum (upper limit) for the scale for each parameter, and then normalize the results and apply a weighting factor, in order to combine them into a single total score. The lower limits, upper limits and weighting factors we chose are shown in Table 4, next to each category header: The first number is the weight, the two numbers between parentheses are the lower and upper limits. The final score for each architecture was calculated from the raw scores, limits, and weights for each parameter p as follows:

$$FinalScore = \sum_{p=1}^n Weight_p \times \min\left(\max\left(\frac{RawScore_p - LowerLimit_p}{UpperLimit_p - LowerLimit_p}, 0\right), 1\right) \quad (1)$$

The minimum possible score is 0 and the maximum possible score is the sum of the parameter weights, which is 46.

Lower limits were determined by considering the score that would correspond to a minimally acceptable cardiac scanner; upper limits correspond to the maximum performance that would provide useful benefit in a cardiac scanner (i.e. architectures were not rewarded for providing performance beyond practical value). For example, for Temporal resolution capability, the lower limit of 0 corresponds to a contemporary single-source CT system with a gantry rotation time of approximately 300 ms and temporal resolution of 150 ms (three times lower than our target performance); the upper limit of 4 corresponds to a temporal resolution of approximately 10 ms (five times as good as our target performance).

The weighting factors were determined as follows. The Value parameter was assigned a weight approximately equal to the average of the Performance parameters, because we decided that for any linear cost/performance relationship, an incremental improvement in performance would correlate to a willingness to incur a proportional incremental cost increase. Parameters within the Performance category were weighted according to the relative importance of each, based on expert consensus. Parameters in the Simplicity category were given relatively low weights, since we felt that if warranted by Performance opportunities, commercial vendors would be willing to invest in developing the required technology. FFOV capability was considered to be a relatively unimportant feature in a cardiac-specific scanner, since these scanners would typically be expensive, installed in specialized heart centers, and would be unlikely to be used for general-purpose scanning.

3. Results and discussion

The results of our scored comparison are shown in Table 3: Raw scores and Table 5: Weighted normalized scores. The final sums of the weighted raw scores are plotted and shown in Fig. 10. Figure 11 shows sketches of all candidate systems, with all preferred architectures outlined; these are also bolded in Table 3 and Table 5.

We reviewed the compiled results with a large group, including two cardiologists, and in several follow-up group discussions. Our decision to use a scoring system to quantify our qualitative assessment of these architectures proved to be a very valuable exercise – it had the effect of guiding our discussions and evaluation toward rigorous, supportable conclusions. However, it did not illuminate any non-intuitive conclusions, and the results show that several of the architectures we considered scored quite similarly, after all the important tradeoffs were considered. These similar scores were the result of tradeoffs balancing the strengths and weaknesses of each candidate. For example, more cost and complexity was offset by higher performance, and more simplicity was offset by lower performance. The strengths and weaknesses of each candidate can be seen clearly in the parameters' sub-scores. In this section, we discuss the down-selection process of the few architectures we will be considering for further evaluation.

Given that this is still a high-level study, we intend to select a small number of representative systems that have the *highest* scores within their category and have a *high* overall score compared to all systems.

In general, CT system design comes down to a few important tradeoffs, including performance versus cost, and IQ versus patient dose. These are the most relevant tradeoffs that were identified by our study.

The first group, “Conventional source, conventional detector architectures”, demonstrates the result of trading off temporal resolution and cost by varying the number of source-detector pairs.

SSCT-FFOV turned out to have a score that is the highest score within its category and high overall, which is not surprising considering that after more than 30 years of development by thousands of scientists and engineers in industry and academia, all four major CT system manufacturers offer products that are evolving toward this architecture. Clearly, SSCT-FFOV should be considered for further analysis.

DSCT-MFOV also received a score that is the highest score within its category and high overall and it is related to a commercially available system, so therefore this architecture should be considered for further comparative studies.

TSCT and QSCT architectures received high scores in general and TSCT-CFOV is the highest within this combined category. Although conceptually appealing, QSCT offers no measurable advantage relative to TSCT that would justify its high cost and complexity, especially since it’s impractical without revision to the geometry.

The next three groups (“Twin Z source, conventional detector architectures”, “Linear source, conventional detector architectures”, and “2D source (Inverse geometry) architectures”) all attempt to trade off cost versus IQ and dose, by varying the number of sources in the longitudinal direction. The IGCT architectures also use multiple sources in the transaxial direction. This results in depositing all radiation dose within the VOI, for improved dose efficiency. In addition, they each can trade off cost versus temporal resolution through the number of source-detector pairs.

Among the TZCT and LSCT variants, the scores were still reasonably high, with the TZCT-FFOV receiving the highest score. The LSCT architectures are an extension of the TZCT systems, but don’t offer sufficient benefit to justify the additional cost and complexity of the required yet-to-be-developed source technology.

IGCT still scored reasonably well and provides a unique dose-reduction capability (“electronic bowtie”), so we selected the highest scoring IGCT-CFOV for further evaluation.

The next group, “Fully stationary CT architectures”, attempts to maximize temporal resolution by using only stationary (non-rotating) components. Unfortunately, all of the architectures that we can imagine have a fundamental shortcoming in data completeness and the cost of these systems would be very high, as reflected by the low scores, so we did not select any of them for further evaluation.

The “Semi-stationary architectures” attempt to resolve these limitations by using one rotating and one stationary component. Although most did not score exceptionally well,

OSDRDCT-CFOV stood out as the preferred alternative, since it can clearly meet the temporal resolution and IQ requirements.

Finally, the novel architectures had lower scores overall. CSCT-CFOV is based on an arguably unsupportable assumption: Spatial repeatability of heart anatomy over multiple cardiac cycles. CCCT-CFOV is exceedingly complex, and to even conceive of implementation, one would need to reduce gantry speed and compromise temporal resolution, and also use an ultra-light source and detector and compromise noise and dose efficiency. Due to their mediocre scores and fundamental shortcomings, we excluded these architectures from further consideration in future work.

4. Conclusion

We followed a systematic process to identify, analyze, and quantitatively compare a broad range of CT system architectures for cardiac-specific CT scanning. We recruited a large team of fifteen scientists, engineers, and cardiologists to participate in this process, and we consulted with more experts in the fields of x-ray sources, detectors, rotating gantries, and image reconstruction. We started by identifying the key clinical applications and performance criteria required of a cardiac-specific CT scanner by performing a literature search and consulting with clinical experts. We identified thirty potential architectures, and documented these with sketches and descriptions of their operation. We objectively rated their relative merits for nine parameters in the categories of Value, Performance, Simplicity, and FFOV capability. We then reviewed these results, and chose six architectures for further, more detailed analysis and comparison, in future work.

The architectures we selected for further evaluation are bolded in Table 5: SSCT-FFOV, DSCT-MFOV, TSCT-CFOV, TZCT-FFOV, IGCT-FFOV and OSDRDCT-CFOV.

Our intention was not to make a competitive comparison (none of the systems exactly match commercial systems), but to give an overview of the different tradeoffs that are important in selecting cardiac CT architectures and provide some expert opinion on what the most promising directions are for cardiac CT.

In future work, we will develop methods to more quantitatively and thoroughly analyze and compare the selected architectures, and identify the one that is most promising for a cardiac-specific CT scanner design. The selected architectures can be further adjusted as we do more research, so we will not limit ourselves to the variations and design details described in this paper.

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Fig. 2.

Single source-detector pair architectures. (a) Single-source CT, Full FOV (SSCT-FFOV). A third-generation architecture that employs a single x-ray source and detector combination. Left: Transaxial cross-section at the central longitudinal plane. Right: Longitudinal cross-section through the axis of rotation. (b) Single-source CT, Cardiac FOV (SSCT-CFOV).

Fig. 3. Dual source-detector pair architectures. (a) Dual-source CT, Mixed FOV (DSCT-MFOV). This architecture employs two x-ray sources and two detectors, one that provides FFOV projections and one that provides CFOV projection. (b) Dual-source CT, Cardiac FOV (DSCT-CFOV).

Fig. 4. Multiple source-detector pair architectures. (a) Triple-source CT, Cardiac FOV (TSCT-CFOV). This architecture employs three x-ray sources and detectors, each providing CFOV. (b) Triple-source CT, Mixed FOV (TSCT-MFOV). (c) Triple-source CT, Cardiac FOV, asymmetrical (TSCT-CFOV(A)). (d) Quintuple-source CT, Cardiac FOV (QSCT-CFOV).

Fig. 5.

Architectures with multiple sources in Z. (a) Dual Twin Z CT, Cardiac FOV (DTZCT-CFOV). This architecture employs two x-ray sources and detectors, each providing CFOV, and with each source containing two focal spots in the longitudinal direction, for reduced cone-beam effects. (b) Dual Twin Z CT, Cardiac FOV (DTZCT-CFOV). (c) Dual Linear Source CT, Mixed FOV (DLSCT-MFOV). (d) Dual Linear Source CT, Cardiac FOV (DLSCT-CFOV).

Fig. 6. Inverted geometry architectures. (a) Inverted Geometry CT, Cardiac FOV (IGCT-CFOV). This architecture employs a 2D x-ray source and incorporates a CFOV. This topology permits a substantially smaller detector. (b) Inverted Geometry CT, Full FOV (IGCT-FFOV).

Fig. 7. Stationary CT architectures. (a) Single Peek-hole Source Ring (SCT-SPHSR-CFOV). (b) Single Offset Source Arc (SCT-SOSA-CFOV). (c) Dual Offset Source Ring (SCT-DOSR-CFOV).

Fig. 8.

Semi-stationary architectures. (a) Ring Source, Dual Rotating Detector CT, Cardiac FOV (OSDRDCT-CFOV). This architecture employs a 360° ring x-ray source and two CFOV detectors mounted on a high-speed rotating gantry. This permits higher temporal resolution without the high cost of a full-ring stationary detector. (b) Ring Source, Dual Rotating Detector Inverse Geometry CT, Cardiac FOV (OSRDIGCT-CFOV). (c) Rotating Source, Ring Detector CT, Cardiac FOV (RSODCT-CFOV).

Fig. 9.

Novel architectures: (a) Sector Source CT, Cardiac FOV (CSCT-CFOV). This architecture employs rotating multi-spot x-ray source and rotates slowly in order to obtain projection images over multiple cardiac cycles. This permits very high temporal resolution without the challenges associated with fast gantry rotation. (b) Concentric Circling CT (CCCT-CFOV).

Fig. 10.
Plot of final total of weighted normalized scores.

Fig. 11.
Summary of all Candidate Architectures. The architectures that were eventually selected for further evaluation are indicated with a bold outline.

Table 1

Key clinical requirements for cardiac imaging

Parameter	Value	Units	Conditions
Transaxial Field of View (FOV)	300	millimeters (mm)	Over the entire Longitudinal coverage
Longitudinal coverage	160	mm	At axis of rotation
Temporal resolution	50	milliseconds (ms)	Nominally, at the center of the FOV
Spatial resolution (isotropic)	20	Line pairs/centimeter (lp/cm)	At isocenter (center of imaged volume), at modulation transfer function (MTF) = 5%
Image quality (noise)	10	Hounsfield Units (HU)	Standard deviation (after noise-reducing image reconstruction algorithms are applied) in a uniform region in the mediastinum of a "typical" size patient
Image quality (artifacts)	None		No artifacts that degrade diagnostic quality
Radiation dose	<1	mSv	"Effective dose"

Table 2

Constant design values for system comparison

Parameter	Value	Units	Conditions
Gantry rotation period (minimum)	150	ms	For stationary-source, rotating-detector architectures
	200	ms	For all other rotating architectures
Source to isocenter distance (SID)	540	mm	From focal spot at anode to axis of rotation
Detector to isocenter distance (DID)	410	mm	From surface of detector to axis of rotation
Focal spot size	0.5	mm	Full Width at Half Maximum (FWHM) of focal spot intensity profile, reflected at iso-center
Detector pixel size	0.5	mm	Isotropic detector pixel pitch, at iso-center
X-ray source power	100	kW	Rotating anode sources
X-ray source power	50	kW	Stationary anode “ring” sources and “IGCT” sources

Table 3

Raw scores

Acronym	Description	Value	Performance				Simplicity			Additional capability
		Total System Cost	IQ - artifacts - raw data completeness	IQ - noise - x-ray fluence capability	Temporal resolution capability	Dose efficiency	System simplicity	Calibration simplicity	Reconstruction simplicity	FFOV capability
SSCT-FFOV	single source CT, full FOV	1.00	5	7	1	7	8	5	5	10
SSCT-CFOV	single source CT, cardiac FOV	0.71	5	7	1	7	9	4	5	0
DSCT-FFOV	dual source CT, full FOV	1.88	5	7	2	6	6	3	5	10
DSCT-MFOV	dual source CT, mixed FOV	1.59	5	7	2	6	7	2	5	10
DSCT-CFOV	dual source CT, cardiac FOV	1.30	5	7	2	6	7	2	5	0
TSCT-FFOV	triple source CT, full FOV	2.75	5	9	3	6	1	2	5	10
TSCT-MFOV	triple source CT, mixed FOV	2.18	5	9	3	6	2	1	5	10
TSCT-CFOV	triple source CT, cardiac FOV	1.89	5	9	3	6	6	1	5	0
TSCT-CFOV (A)	triple source CT, cardiac FOV (asymmetrical)	1.89	5	7	3	6	2	1	5	0
QSCT-CFOV	quintuple source CT, cardiac FOV	2.67	5	9	4	5	2	0	5	0
TZCT-FFOV	twin Z CT, full FOV	0.94	6	5	1	7	7	4	5	10
TZCT-CFOV	twin Z CT, cardiac FOV	0.69	6	5	1	7	8	3	5	0
DTZCT-FFOV	dual twin Z CT, full FOV	1.69	6	5	2	6	5	2	5	10
DTZCT-MFOV	dual twin Z CT, mixed FOV	1.44	6	5	2	6	6	1	5	10
DTZCT-CFOV	dual twin Z CT, cardiac FOV	1.19	6	5	2	6	6	1	5	0
LSCT-FFOV	linear source CT, full FOV	0.83	6	5	1	7	3	4	5	10
LSCT-CFOV	linear source CT, cardiac FOV	0.66	6	5	1	7	4	3	5	0
DLST-MFOV	dual linear source CT, mixed FOV	1.11	6	5	2	6	2	1	5	10

Acronym	Description	Value	Performance				Simplicity			Additional capability
		Total System Cost	IQ - artifacts - raw data completeness	IQ - noise - x-ray fluence capability	Temporal resolution capability	Dose efficiency	System simplicity	Calibration simplicity	Reconstruction simplicity	FFOV capability
DLSCT-CFOV	dual linear source CT, cardiac FOV	0.95	6	5	2	6	2	1	5	0
IGCT-FFOV	inverted geometry CT, full FOV	0.53	6	2	1	7	4	2	5	10
IGCT-CFOV	inverted geometry CT, cardiac FOV	0.47	6	2	1	7	5	1	5	0
SCT-SPHSR-CFOV	stationary CT, single "peek-hole" source ring, cardiac FOV	2.92	1	3	2	7	3	3	4	10
SCT-SOSA-CFOV	stationary CT, single offset source arc, cardiac FOV	2.46	4	3	2	7	3	3	5	0
SCT-DOSR-CFOV	stationary CT, dual offset source ring, cardiac FOV	2.36	3	3	2	2	3	3	4	10
OSRDCT-CFOV	ring source, rotating detector CT, cardiac FOV	1.70	5	4	2	7	4	2	5	0
OSDRDCT-CFOV	ring source, dual rotating detector CT, cardiac FOV	2.13	5	6	3	6	2	0	5	0
OSRDIGCT-CFOV	ring source, rotating detector inverse geometry CT, cardiac FOV	1.41	5	2	1	7	4	0	5	0
RSODCT-CFOV	rotating source, ring detector CT, cardiac FOV	3.57	5	7	1	7	6	5	5	10
CSCT-CFOV	sector source CT, cardiac FOV	0.95	5	7	1	2	5	2	4	5
CCCT-CFOV	composite circling CT, cardiac FOV	1.62	6	3	1	7	1	2	4	0

Table 4

System evaluation parameters and scoring system

Performance		Simplicity	
IQ - artifacts - completeness	7 (1,7)	System simplicity	4 (0,10)
TZCT, LSCT, IGCT, CCCT	6	Default	8
Default	5	TZCT	7
SCT-SOSA	4	DSCT, RSODCT	6
SCT-DOSR	3	TSCT-CFOV, DTZCT	5
SCT-SPHSR	1	IGCT-FFOV, CSCT	4
		LSCT, SCT-*, OSRD	3
IQ - noise - x-ray fluence	7 (2,10)	OSDRD, OSIGCT	2
TSCT, QSCT	9	TSCT-FFOV, TSCT(A), QSCT, DLSCT	1
Default	7	CCCT	0
OSDRD	6	Add for 1 CFOV detector	0.5
TZCT, LSCT	5	Add for all CFOV detectors	1
OSRD	4		
SCT-*, CCCT	3	Calibration simplicity	1 (0,6)
IGCT, OSRDIGCT	2	Default	5
		TZCT, LSCT, CCCT	4
Temporal resolution	10 (0,4)	DSCT, SCT-*, OSRDCT, CSCT	3
QSCT	4	TSCT, TSCT(A), DTZCT, DLSCT, IGCT	2
TSCT, TSCT(A), OSDRD	3	QSCT, OSDRD, OSRDIGCT	1
DSCT, DTZCT, DLSCT, SCT-*	2	CCCT	0
OSRD	1.5	Subtract if any CFOV detectors	1
Default	1		
CSCT	0	Reconstruction simplicity	1 (2,7)
		Default	5
Dose efficiency	7 (0,9)	SCT-SPHSR, SCT-DOSR, CSCT, CCCT	4
Default	7		
DSCT	6	FFOV	1 (0,10)
TSCT, TSCT(A)	5.5	Yes	10
QSCT	5	CSCT	5
SCT-DOSR, CSCT	2	No	0

Table 5

Weighted normalized scores

Acronym	Description	Total weighted normalized score	Value	Performance	IQ - artifacts - raw data completeness	IQ - noise - x-ray fluence capability	Temporal resolution capability	Dose efficiency	Simplicity	System simplicity	Calibration simplicity
SSCT-FFOV	single source CT, full FOV	28	5	17	5	4	3	5	5	3	1
SSCT-CFOV	single source CT, cardiac FOV	28	6	17	5	4	3	5	5	4	1
DSCT-FFOV	dual source CT, full FOV	27	3	19	5	4	5	5	4	2	1
DSCT-MFOV	dual source CT, mixed FOV	27	4	19	5	4	5	5	4	3	0
DSCT-CFOV	dual source CT, cardiac FOV	27	5	19	5	4	5	5	4	3	0
TSCT-FFOV	triple source CT, full FOV	27	2	23	5	6	8	4	1	0	0
TSCT-MFOV	triple source CT, mixed FOV	28	3	23	5	6	8	4	1	1	0
TSCT-CFOV	triple source CT, cardiac FOV	29	3	23	5	6	8	4	3	2	0
TSCT-CFOV (A)	triple source CT, cardiac FOV (asymmetrical)	26	3	21	5	4	8	4	2	1	0
QSCT-CFOV	quintuple source CT, cardiac FOV	28	2	25	5	6	10	4	1	1	0
TZCT-FFOV	twin Z CT, full FOV	27	5	16	6	3	3	5	4	3	1
TZCT-CFOV	twin Z CT, cardiac FOV	26	6	16	6	3	3	5	4	3	1
DTZCT-FFOV	dual twin Z CT, full FOV	25	4	18	6	3	5	4	3	2	0
DTZCT-MFOV	dual twin Z CT, mixed FOV	26	4	18	6	3	5	4	3	2	0
DTZCT-CFOV	dual twin Z CT, cardiac FOV	26	5	18	6	3	5	4	3	2	0
LSCT-FFOV	linear source CT, full FOV	25	5	16	6	3	3	5	2	1	1
LSCT-CFOV	linear source CT, cardiac FOV	25	6	16	6	3	3	5	3	2	1
DLSTCT-MFOV	dual linear source CT, mixed FOV	25	5	18	6	3	5	4	1	1	0
DLSTCT-CFOV	dual linear source CT, cardiac FOV	25	5	18	6	3	5	4	2	1	0

Acronym	Description	Total weighted normalized score	Value	Performance	IQ - artifacts - raw data completeness	IQ - noise - x-ray fluence capability	Temporal resolution capability	Dose efficiency	Simplicity	System simplicity	Calibration simplicity
IGCT-FFOV	inverted geometry CT, full FOV	23	6	14	6	0	3	5	3	2	0
IGCT-CFOV	inverted geometry CT, cardiac FOV	23	6	14	6	0	3	5	3	2	0
SCT-SPHSR-CFOV	stationary CT, single “peek-hole” source ring, cardiac FOV	16	1	11	0	1	5	5	2	1	1
SCT-SOSA-CFOV	stationary CT, single offset source arc, cardiac FOV	19	2	15	4	1	5	5	2	1	1
SCT-DOSR-CFOV	stationary CT, dual offset source ring, cardiac FOV	15	2	10	2	1	5	2	2	1	1
OSRDCT-CFOV	ring source, rotating detector CT, cardiac FOV	22	4	16	5	2	4	5	3	2	0
OSDRDCT-CFOV	ring source, dual rotating detector CT, cardiac FOV	24	3	20	5	4	8	4	1	1	0
OSRDIGCT-CFOV	ring source, rotating detector inverse geometry CT, cardiac FOV	19	4	13	5	0	3	5	2	2	0
RSODCT-CFOV	rotating source, ring detector CT, cardiac FOV	22	0	17	5	4	3	5	4	2	1
CSCT-CFOV	sector source CT, cardiac FOV	22	5	13	5	4	3	2	3	2	0
CCCT-CFOV	composite circling CT, cardiac FOV	20	4	15	6	1	3	5	1	0	0