

Maximizing dose reductions with cardiac CT

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Abstract Multidetector computed tomography has come a long way in a short time, quickly becoming a standard tool in the cardiac imaging armamentarium. The promise of plaque imaging, combined with both anatomical visualization and stenosis detection, has made this a preferred first line test of many cardiologists and radiologists. This test is well suited to rule out coronary artery disease (obstruction) and still diagnosing subclinical plaque, with may be a good target for anti-atherosclerotic therapies. There has been recent criticism against CT imaging, and cardiac CT specifically, due to the high radiation doses that being employed. New advances have allowed for dramatic dose reductions. These include more routinely performed methods such as dose modulation, and newer methods such as prospective gating or minimizing the field of view. This paper will review the different applications to reduce cardiac CT radiation doses to nominal levels, potentially expanding the applications of cardiac CT by removing one of the biggest barriers.

Keywords Radiation exposure · Radiation reduction · Millisievert · Noninvasive angiography · CT angiography, CTA · Computed tomographic

angiography · Multidetector computed tomography, MDCT · Multislice computed tomography, MSCT

Introduction

Over the last decade great strides have been made in the field of cardiac imaging, particularly in the ability to view the coronary artery lumen with ample diagnostic accuracy [1]. Current 64-MDCT (multi-row detector computed tomography) systems have faster gantry rotation speed resulting in better temporal resolution. Better Z-axis spatial resolution is made possible by thin collimations with extensive volumetric acquisitions [2, 3]. The technique of retrospective image acquisition scanning with overlapping of slices that compensates for potential cardiac motion data at all phases of the cardiac cycle to assess ejection fraction valves and wall motion. [2, 3]. This results in more patient radiation exposure in comparison with CTA with prospective-gated acquisition (originally used with the electron beam tomography scanner) [4]. Studies estimate radiation exposure for 16-row at 8.8 mSv for a 16×0.75 mm scan protocol with a pitch of 0.28 and power of 370 mA [5], and 13 and 18 mSv (for men and women, respectively) with 64-row MDCT [6]. In a recent study, Pugliese et al. [7] reported radiation doses in the order of 15:20 mSv (male:female),

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respectively. These doses for retrospective imaging are significantly higher than those seen with invasive diagnostic coronary angiography where the dose is between 2.1 and 7 mSv [8, 9], but lower than the effective radiation dose for a technetium Tc 99m sestamibi myocardial perfusion stress test (9–18 mSv, increasing to over 20 mSv with dual-isotope scanning) [10, 11].

Methods to minimize dose

Multiple different radiation reductions methods have been reported, all of which can result in complementary reductions in radiation exposure. This paper outlines reported and potential methods, as well as discusses the potential for future radiation reduction techniques. While it is logical and in keeping with the principles of ALARA (as low as reasonably achievable) to use the minimum radiation dose necessary in cardiac CT examinations, there are little data testing the effect of specific radiation sparing approaches on diagnostic quality. Cardiac CTA imaging protocols are also not as widely standardized, and thus maintaining diagnostic accuracy is important. An examination which is non-diagnostic, of course, exposes the patient to radiation without benefit. However, experience dictates that through attention to detail of the CT acquisition, excellent image quality can be maintained at a lower radiation exposure.

Dose related issues

Limiting craniocaudal coverage length

The region of the patient's body that undergoes radiation exposure in the craniocaudal, or z-direction is the largest contributor to radiation absorption (via changes in the dose length product [DLP]). For cardiac CT examinations, proper attention to scan length should be made. For example, rather than rigidly scanning from carina to diaphragm as is frequently recommended by scanner manufacturers, the operator could potentially use the non-contrast images for calcium scoring, if necessary to ascertain the most cranial and caudal position of the coronary arteries.

Using the scout image, the field of view usually extends from ~1 cm below the carina to just below the diaphragm to ensure complete coronary imaging (Fig. 1). The scout film is highly inaccurate for determining the precise origin of the coronary arteries [12]. An alternative is to use images from the coronary calcium scoring to set the upper limit above the apex of the left anterior descending artery (the most superior artery) and the lower limit inferior to the posterior descending artery, leaving sufficient but not excessive margins, to allow for movement. Calcium scoring with MDCT delivers ~1.2 mSv when acquired with prospective ECG gating, and potentially lower if 100 kVp is used for this acquisition [4]. A

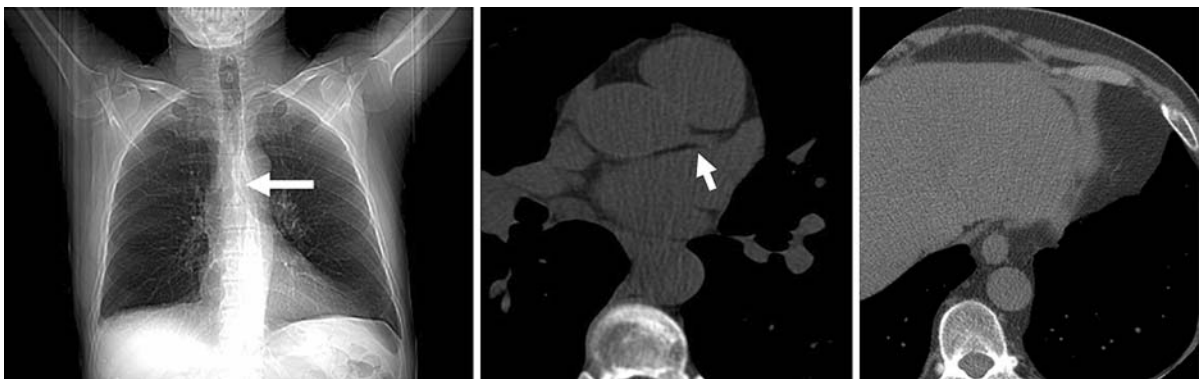


Fig. 1 A scout film (*left panel*) used to identify the upper and the lower limits of the region of interest for the subsequent CTA. The *white arrow* shows the level of the carina, which is a landmark for determining the top of the coronary tree. *Middle panel* An axial slice of a non-contrast coronary calcium scan showing the left main artery (*black arrow*). The exact level in which the superior most coronary arteries arise is most

effectively visualized with axial data such as the calcium scan. *Right panel* An axial slice of a non-contrast coronary calcium scan defining the lower limits of the heart, demonstrating the level just below the posterior descending artery. There are no coronary arteries on this image, so concluding the CTA just below this would safely allow for complete imaging of the coronary tree

30% reduction in radiation dose has been shown by employing a calcium score to determine the superior and inferior borders of the field of view, with the superior border at 1 cm above the visualized top of the coronary arteries and the inferior border at 1 cm below the posterior descending artery [13]. The savings in radiation dose was 4 mSv offsetting the dose delivered by calcium scoring (1.2 mSv). The average percentage reduction in radiation was 30% (SD \pm 10) with a median of 22%. This method of dose reduction is incremental to dose modulation as well as reducing the mA or kV during image acquisition. Multipurpose examinations of the chest have long Z axis acquisitions and thus higher radiation dose. Multipurpose examinations of the chest tend to have long Z axis acquisitions and very high radiation exposures. For example, to do a “triple rule out” examination, both scan length and scan time are affected, and there is a consequent marked increase in radiation exposure. Thus, this protocol should be clinically indicated to justify the additional radiation.

Tube amperage

Tube mA is a variable setting that affects the number of photons generated and directly affects image signal to noise ratio. Radiation dose is approximately proportional to mA, and customarily, amperage can be adjusted for body mass and configuration [3]. Patients with higher mass will experience higher photon scatter and higher noise, while thinner patient will have images of good quality using lower amperage. Failure to adjust mA downward for thin patients will result in unnecessary radiation. It is important to reduce mA to the lowest necessary to make diagnostic images. Typical doses are 350 mA for small patients, 450 mA for medium sized patients and 550 or higher for large patients (scaled to maximum mA of the X-ray tube for the heaviest patients). Typical settings from the manufacturers are pre-set higher, to optimize image quality at the expense of radiation dose to the patient.

Dose modulation (electrocardiographic pulsing)

Current generation scanners are capable of varying tube current output (mA) in synchrony to the patient’s electrocardiogram. This is done to reduce radiation during phases in the cardiac cycle when the heart is

moving more dynamically (ventricular systole and end diastole [which correlates with atrial systole]). The ideal “pulsing window” (when current becomes maximal) is as short as possible, typically focused around the 70% phase of the cardiac cycle. This becomes a complex decision as there is a trade-off between pulse window width, heart rate and scanner type. All current-generation cardiac-capable scanners have built-in software that either varies pulsing window width with heart rate or allows the operator to customize these protocols. The use of ECG pulsing can decrease radiation dose by 50% or more and is generally recommended unless other parameters threaten the image quality (such as irregular heart rate). The time of least motion occurs between 40 and 80% of the R–R interval (mid diastole) [14, 15]. With dose modulation, the dose is reduced by 18–47% depending on the patient’s heart rate [16–18]. Slower heart rates will lead to more effective use of dose-modulation and lower radiation doses (with the exception of the dual source scanner, where slower heart rates will increase radiation exposure). The multicenter 64-MDCT ACCURACY trial utilized dose modulation and still reported negative predictive values of 99% [1]. While EKG-regulated dose modulation could be implemented in a majority of patients, sometimes it cannot be utilized due to scanning conditions requiring additional image reconstructions during different phases of the cardiac cycle (such as irregular heart rhythms and fast heart beats). Although the radiation burden of cardiac CT studies can be efficiently reduced by dose modulation, a further decrease in radiation without compromising diagnostic image quality would be indeed very desirable [19].

Scanner type

In general, increasing the number of detector-rows and reducing detector size tends to increase the radiation dose due to the increasing surface area of lead collimators (which can only be so thin while still being effective) in comparison to detector area. The more lateral detectors that are present lead to less efficient delivery of dose, as the lateral detectors (those not directly across from the X-ray tube) require more radiation exposure due to scatter. Thus, four row scanners had lower patient radiation doses than 16 slice scanners, despite longer radiation times.

Similarly 16 row scanners impart lower doses than 64, and subsequently, doses from 64 will be lower than 256 and 320 row scanners (for similar protocols). In addition, complex effects are also produced by dual source scanners which have two X-ray sources and detector rings operating during the scan time, but have a reduced scan time and heart-rate variable pitch. Theoretically, dual source scanners reduce the scan duration because their faster temporal resolution allows imaging faster heart rates thereby reducing diastolic time, which balances the double dose of radiation being emitted, leading to similar doses as 64 row MDCT scanners and lower doses than 256 and 320 row scanners.

Reducing tube voltage

Another methodology to reduce the dose is to reduce tube voltage. Tube kVp affects the peak photon energy and affects image contrast. In general, kVp has been used to frequently adjust for body mass. New studies suggest that that protocols utilizing 100 or 80 kVp may be effective in thin patients for reducing dosage in coronary calcium measurements or coronary CT angiography, without degrading diagnostic accuracy [20]. Tube voltage has a more dramatic effect on radiation dosage, which varies approximately with the square of the kVp. Budoff et al. [21] demonstrated use of 100 kVp reduced radiation dose 42% using prospective triggering and 40% using retrospective imaging, as compared to 120 kVp ($P < 0.001$). Hausleiter et al. [22] showed a 53–64% reduction in estimated radiation dose using both reduced kVp and dose modulation. As noted above, this benefit is independent and incremental to dose modulation radiation savings and other savings listed above (Table 1).

Limiting field of view (Fov)

An important technique that is currently underused, which will both minimize radiation and afford better image quality, is to restrict the xy field of view. A smaller xy field of view will improve image quality, as the FOV divided by 512 is the resolution in the X – Y axis (to an optimal resolution of about 0.3–0.35 mm for current scanners). The bowtie filter allows for smaller radiation exposure by limiting the scatter of the X-ray towards the detectors. Intrinsically, it is more efficient to irradiate detectors directly across from the X-ray tube and thus, requires less photons. Thus, the bowtie filter allows less X-rays out of the central part of the X-ray tube, as less are required to expose the detectors. As one moves laterally from the center of the detector array, the number of photons needed to create an image is higher (less efficiency and thus, the relative dose increases for scanners with higher numbers of detectors (see #4 above). The bowtie filter allows more photons to go toward the lateral detectors, thus looking like a bowtie, with a small center and large lateral edges. These filters are optimized for patient size, and the smallest bowtie filters, only allow the lateral tissue to be exposed to 32 cm, the medium bowtie to 36 cm and the largest bowtie to over 40 cm. Since all hearts (including bypass patients) fit in a small bowtie filter, the use of larger bow-ties would only be necessary when larger field of views are needed. In a study of radiation doses (measured radiation exposures using the LightSpeed VCT 64-MDCT scanner (GE Healthcare, Waukesha, WI, USA), the dose using the small cardiac bowtie (can only be post-processed up to 32 cm field of view), resulted in a dose of 8 mSv (retrospective imaging, as low as 1 mSv with prospective imaging) [23]. A medium bowtie filter for a retrospectively acquired

Table 1 Expected dose reductions based upon different scanning strategies

Dose reduction technique	Effectively lowers exposure with retrospective gating	Estimated dose reduction for retrospective	Effectively lowers exposure with prospective triggering
Scan length	↓↓↓	Proportional	↓
Adding calcium Score to lower scan length	↓↓	30%	↑
Reducing amperage (mA)	↓	Proportional	↓
Dose modulation	↓↓↓	20–40%	Not applicable
Reducing tube voltage to 100 kVp	↓↓↓	40%	↓↓↓
Limiting scan field of view	↓↓↓	67%	↓↓

↓—Mild decrease; ↓↓—moderate decrease; ↓↓↓—large decrease in radiation dose

cardiac CT angiogram using the same protocol affords 13.4 mSv of radiation. This is a 40% dose reduction by simply using an appropriate sized bowtie filter for the study required (cardiac CT requires at most 25 cm of view). Larger patients may require a medium bowtie filter due to optimization protocols, but for cardiac applications, a large bowtie filter would never be necessary. The routine use of small bowtie filters (and small FOV) will both reduce radiation dose and limit incidental findings related to non-cardiac disease (primarily non-cancerous lung nodules). There is no evidence yet to support that discovering these incidental findings will lead to improvement in outcomes. There is data available that these incidental findings increase surgical rates, costs and patient anxiety [24]. For recommended bowtie filters based on body habitus for the GE VCT Scanner, see Table 2. These findings need to be validated for other scanners and manufacturers.

Sequential (prospective) scanning versus retrospective gating

Until recently, all studies were done with retrospective gating, using continuous X-ray beam exposure to create thousands of images, then selecting a subset of those at the optimal phase of the cardiac cycle for interpretation. An alternative imaging mode is the axial step-and-shoot acquisition (prospective gating). This was first used with electron beam CT angiography as early as 1995, and with calcium scoring with MDCT since 1998. Approximately 2 years ago,

prospective gating was introduced for MDCT angiography. This is by far the most significant dose reduction technique, as it turns off the X-ray tube except for a short exposure period centered around the ideal imaging phase (typically mid-diastole). In patients with slow and very steady heart rates, extremely low radiation doses can be achieved by sequential scanning with tube output only during a narrow ECG window (Table 3). The X-ray beam is triggered by the ECG and is turned on only during the required phase of the cardiac cycle (usually 75% of the cardiac cycle) and turned off during the rest of the cardiac cycle. Retrospective ECG gating can be done with or without ‘padding’. Padding provides additional phase information to account for variations in heart rate by adding time before and after the center phase of the acquisition (e.g. adding 100 ms before and after the 75% phase thus providing additional phase information). Using the GE technology, padding is prescribed in the range of 0–200 ms and is added to both sides of the center of the acquisition. This allows more phases of data to be available to the reader and is generally used when heart rates are >60 bpm or when heart rate variability is noted. The padding should be wider with more rapid heart rates [25]. When padding was used, it is added in a heart rate dependent fashion as follows: 30–39 bpm, 175 ms padding; 40–49 bpm, 150 ms padding; 50–59 bpm, 125 ms padding and ≥ 60 bpm, 100 ms padding. Padding default is set to approximately $\pm 10\%$ of the patient’s R–R cycle. Patients with very stable and slow heart rates underwent no padding to

Table 2 Radiation doses of retrospective imaging with 64 MDCT based on body mass index (BMI)

BMI	Calcium scan		Retrospective		KvP	DLP	Estimated mSv including CAC scan		
	kVp	mA	Max mA	Min mA					
19 and less	250	120	300	280	140	100	C2/Small filte	244	4.148
20–22	275	120	350	325	170	100	C2/Small filte	289	4.913
23–25	300	120	375	350	190	100	C2/Small filte	311	5.287
26–28	325	120	450	425	215	100	C2/Small filte	359	6.103
29–31	350	120	500	475	240	120	C2/Small filte	640	10.88
32–35	375	120	600	575	290	120	C2/Small filte	772	13.124
36–37	400	120	700	650	290	120	C3/Med filter	954	16.218
≥ 38	425	120	800	750	375	120	C3/Med filter	1,170	19.89
>40	450	120	800	775	388	120	C3/Med filter	1,270	21.59

DLP dose length product; *kVp* killivolts; *mA* milliamperes; *CAC* coronary artery calcium scan

Table 3 SnapShot Pulse (prospective acquisition) GE-VCT parameters

Action	Parameter
Cardiac mode	Prospective step-and-shoot (SnapShot Pulse)
Rotation time	0.35 s
Start location and end location	Based on scout coverage—base of the heart to apex (about 1 cm above the left main artery to about 1 cm below the posterior descending artery)
Detector coverage	40 mm
Slice thickness	0.625 mm
Padding	Set by system or turned off
SFOV (scan field of view)	Small or medium depending on the body habitus of the patient
DFOV	Typically 18 or 15 cm field of view to include all of the cardiac structures
kVp	120 kVp \geq 85 kg patient (obese or morbidly obese) 100 kVp for < 85 kg patient
mA	350—Small patient 450—Medium patient 550+ —Large patient

Adapted from Gopal et al. [13]

ensure the lowest possible radiation dose. The prescribed PA window is defined as center phase \pm milliseconds padding (Fig. 2).

Multiple studies have demonstrated similar or superior image quality with prospective imaging (LightSpeed VCT 64-MDCT scanner), and resulted in average doses of 1–3 mSv [26–28]. Superior image

quality with prospective imaging is most likely due to lack of ‘Z axis’ motion, as there is no movement of the table (zero pitch) during image acquisition. Multiple studies have reported mean patient radiation doses 77–80% lower for prospective gating than for retrospective gating, without compromising image quality or diagnostic accuracy. Budoff et al. [21] showed a 90% reduction in estimated radiation dose using a combination of reduced kVp (100) and prospective triggering together. In a majority of patients, patients can undergo prospective triggering with marked radiation dose reduction, if ejection fraction information or multiple phase acquisition is not necessary. This removes one of the biggest barriers to more widespread implementation of cardiac CT [29].

Future horizons for dose reduction

The newly released High Definition CT (CT750 HD, GE Healthcare, Waukesha, WI, USA) uses the first new detector material in 20 years. This detector material, by changing the molecular structure of real garnets, a scintillator that is proposed to deliver images 100 times faster, with up to 33% greater detail through the body and up to 47% greater detail in the heart. This also allows up to 2,496 views per rotation (a 2.5 \times increase) to deliver improved spatial resolution and improved image quality across the entire field of view. While validation studies are now underway, the vendor reports a reduced dose by as much as 83% for cardiac CT. The exact application

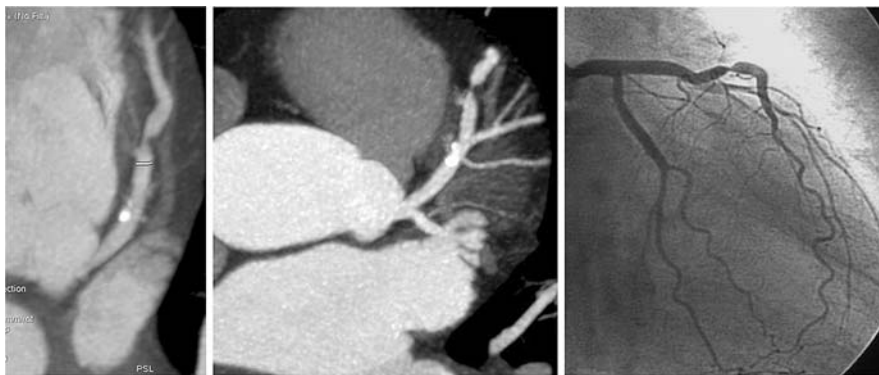


Fig. 2 Shows stenosis in left anterior descending from patient who underwent both cardiac gated helical (RS-OHA) and SnapShot Pulse acquisitions (PA) within 1 week. Images courtesy of Cardiovascular Medical Group, Beverly Hills,

CA. *Left panel* is retrospective image, *middle panel* is prospective image and *right panel* is corresponding cardiac catheterization

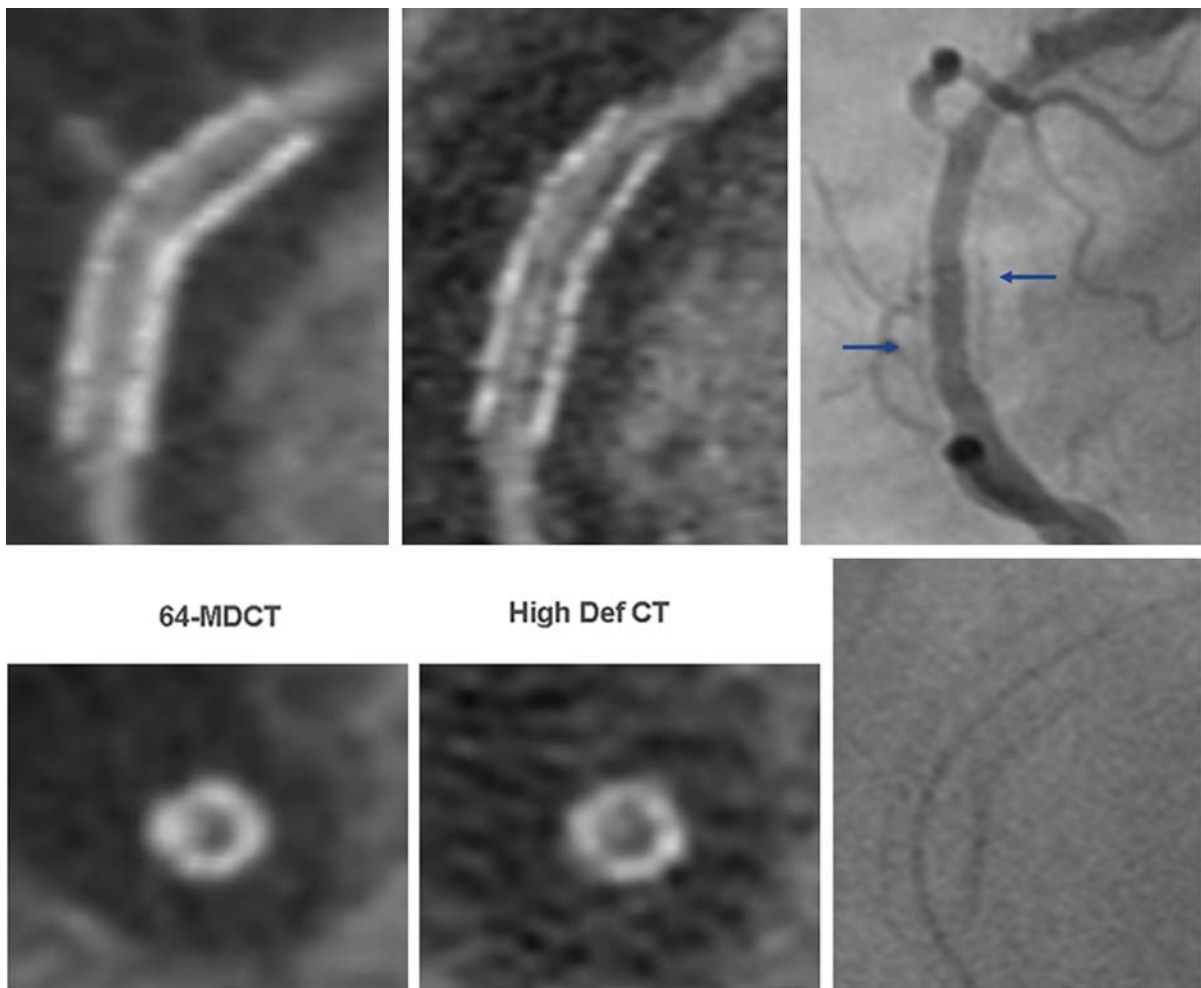


Fig. 3 *Left image* Resolution of stent impaired with current 64-multi-detector computed tomography. *Middle image* Improved visualization of in-stent restenosis by High Definition CT, with confirmation of in-stent restenosis by invasive

angiography. The *arrows* represent the stent borders on angiography. Images courtesy of Dr. J. L. Sablayrolles, Centre Cardiologique du Nord, Saint Denis, France

and overlap with other dose reduction techniques remains to be studied, but the use of more efficient detectors may reduce signal to noise ratios without requiring the typical increase in radiation dose. The High Definition CT should improve visualization of stents and other small structures currently limited by spatial resolution in current 64–320 scanners (Fig. 3).

Summary

Cardiac CT imaging protocols should be constructed to minimize the radiation exposure to the patient

while providing the best possible information for an accurate diagnosis. Diagnostic imaging most likely saves thousands of lives each year by providing medical information to the physician for better medical management of the patient for the overall outcome of the patient's health. While there may be a finite risk associated with a diagnostic study using low level ionizing radiation, the typical patient presents a higher risk by not having the imaging study performed. Using the techniques outlined for adequate patients (based upon body habitus, heart rate and cardiac rhythm), dose reductions up to 70–90% can be achieved when compared to earlier acquisition strategies for cardiac CT.

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