Cardiac CTA, Radiation, and Radiation Safety: Principles and Practice

John A. Rumberger, PhD, MD, FACC and David Fein, MD

Cardiac CTA Training Associates www.cardiaccta.us, and

The Princeton Longevity Center, Princeton, NJ. www.thplc.net

In our continuing education CCTA newsletters we address the complicated issues of Cardiac CTA. In the series to follow we discuss patient radiation dose and radiation safety for Cardiac CTA; understanding these aspects is important for patient care and is a major component of the yearly administered Cardiac CT Board Examination.

Overview: Cardiac CTA and Radiation Safety

The following principles, along with all aspects of CT physics, are derived from our SCCT Certified (www.scct.org), intensive and highly lauded Level II/III Cardiac CT training programs for Cardiologists, Nuclear Medicine Specialists, and Radiologists. The current discussion as presented in this continuing series will be divided into 4 separate parts –

1. How Cardiac CT Works
2. Measuring Radiation Dose in Cardiac CT
3. Principles of ALARA and Radiation Safety in Cardiac CT
4. Methods to Reduce Effective Radiation Dose From Cardiac CT

Part I: How Cardiac CT Works

In general multi-detector cardiac CT and standardized multi-detector CT (MDCT) of the body involve the same structural components; it is just the unique manner in which Cardiac CT must be performed that separates the protocols.

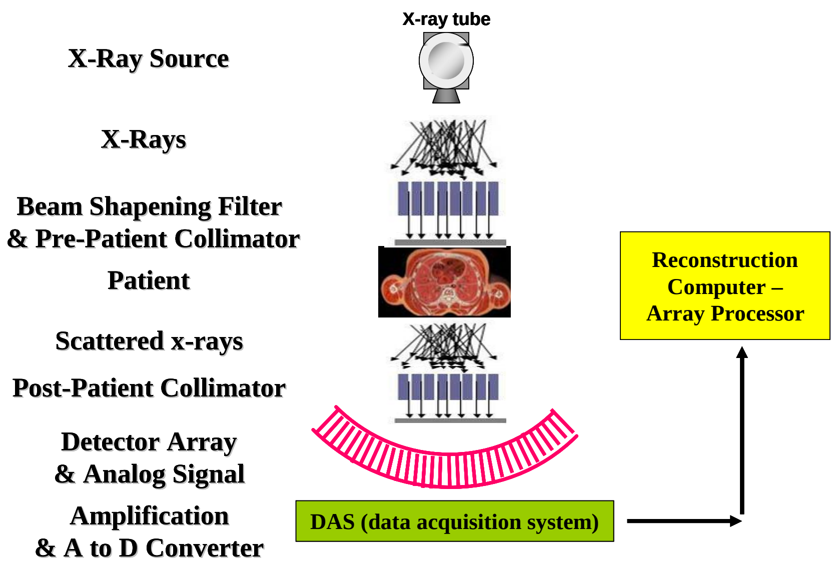


Figure 1

Figure 1 shows the general components of a standard MDCT scanner. The apparatus sits on a gantry and operates independent of the movement of the patient couch. The introduction of 'slip ring' technology in the 1990's allows free spinning of the x-ray tube about the gantry without the need to 'rewind'; this was a necessary step in the introduction of spiral or helical CT scanning.

X-rays are ‘polychromatic’ (multi-energetic) and fan outwardly from the source (x-ray tube). It is most optimal to ‘focus’ these various x-ray energies to be incident upon the body in as parallel a manner as possible and thus, prior to entering the patient, the x-rays are ‘shaped’ and pass through a pre-patient collimator (focuser). When the CT technologist selects a section thickness he or she is determining tube collimation by narrowing or widening the beam.

After the x-rays pass through the patient and are attenuated (ideally) in proportion to the specific attenuation coefficients (a property or characteristic of the tissue) they are again ‘scattered’ and most be ‘re-focused’ using a ‘post-patient’ collimator. The primary responsibility of this set of collimators is to insure proper beam width at the detector to reduce the number of scattered photons which enter, thus reducing noise in the subsequent reconstructed tomograms. These focused photons are incident on an array of specialized ‘detectors’.

A CT detector is a crystal or a ceramic surrounding ionized gas that when struck by an x-ray photon produces light or an electrical signal. The two types of detectors used in CT are the scintillation/solid state detectors and the xenon gas detectors. Crystal solid state detectors were the standard for many years but have been replaced in 64-slice CT with the gas detector. This gas detector is constructed utilizing a chamber made of a ceramic material imbedded with long thin ionization plates usually made of tungsten and submersed or surrounded by Xenon gas. These plates act as electron collecting systems and when photons interact with the xenon gas ionization occurs at the plate, which produces a small electrical current. The term 'detector' refers to a single element but the term 'detector array' is used to describe the total number of detectors that a CT system uses for collecting attenuated x-ray information.

Although the x-rays incident on the detector array produce small electrical potentials (in a manner analogous to a photo-cell producing electricity from the sun's rays) – these signals must be amplified AND converted to digital information. This accomplishment of analog to digital (A to D) conversion is done in the DAS (data acquisition system).

Finally, the digital transmission signal is fed to the 'array processor', which is the specialized computer used to reconstruct the images representing the spatial organization of the x-ray attenuation coefficients in each CT cross-section or slice.

This manner of producing tomographic cross-sectional images is the same regardless of which part of the body is being scanned; however, for cardiac CT, due to imaging of an object in which sub-millimeter detail is required and which, in addition, is in continuous motion during image acquisition – the scanning protocols differ from that of standard body imaging.

Radiation to the patient from CT is largely dependent on two basic setting – x-ray tube current (called mA or milli-Ampere's) and x-ray tube voltage (called kV or kilo-volts). The mA settings can be varied by the CT technologist and in some systems the kV can also be varied; these settings essentially control the flux of x-ray photons and it is essential to have an adequate photon flux (sort of like adequate light exposure for a simple photograph) to get adequate, diagnostic reconstructed images.

Additional parameters however also affect radiation dose and include slice thickness. For imaging of the chest, slice thicknesses of 5-6 mm are adequate, but for imaging of the coronary arteries, with nominal diameters of 1mm to 5 mm, slice thicknesses (essentially the setting of the width of the collimators or collimation) must be sub-millimeter. In the current generation of 64-slice (and higher) row multi-detector CT scanners, the slice thicknesses (collimations) are on the order of 0.5 mm to 0.7 mm. In order to have adequate photon flux to reconstruct images with these nominal dimensions, the photon flux per centimeter must be increased – which then increases the patient radiation exposure for cardiac CT.

The final parameter affecting radiation dose for cardiac CT is 'pitch'. The general principles of pitch using multi-detector CT are illustrated in Figure 2, demonstrating non-overlapping rotations of the gantry. Pitch is defined as the incrementation of the patient

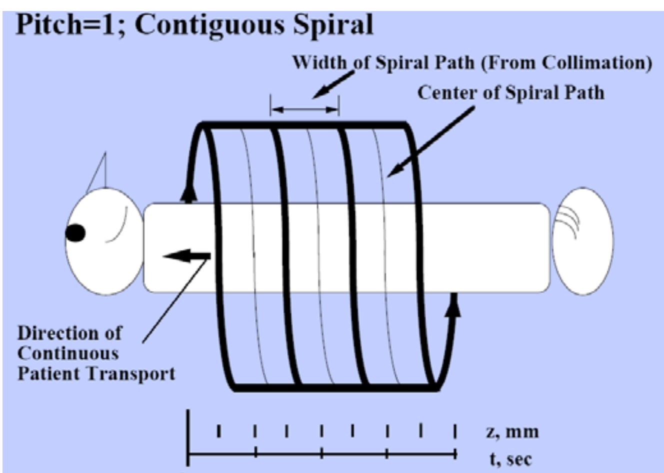


Figure 2:

couch per gantry rotation, divided by the x-ray beam width. For example, say the collimation is set at 0.5 mm (slice thickness) using a 64-slice scanner; this translates into 32 mm of 'coverage' per rotation of the gantry. If the 'pitch' is set at 1.0, the patient couch will move 32 mm for each scanner gantry rotation. For cardiac CT imaging, due to the need to reconstruct images from variable 'phases' during the cardiac cycle, pitch is generally set at 0.5 (16 mm table increment per gantry rotation, for the example given). These 'overlapping' spins of the x-ray camera increase the total radiation dose to the patient. Due to geometric issues, however, the radiation dose for a pitch of 0.5 is not twice that of a pitch of one, but is increased by the square root of 2 (~1.4 x).

In part II we will cover the principles of measuring and determining the 'effective' radiation dose to the patient when performing Cardiac CT.

Part II: Measuring Radiation Dose in Cardiac CT

Part I of this discussion centered on 'How Cardiac CT Works' and commented on the major elements which result in radiation to the patient: x-ray tube current (mA), x-ray tube voltage (kV), slice (collimation) thickness/width, and 'pitch'. In the brief discussion to follow, we will examine how radiation dose using MDCT is determined.

In order to understand CT 'dose' there are some basic definitions that must be put forward and include: 'absorbed dose', 'organ dose', and 'effective (patient) dose'.

Figure 3

A radiation physicist first determines the 'absorbed dose using a tool called the CTDI (CT dosimetry index). This consists of a Plexiglas 'phantom' of various sizes – generally a 17 cm in diameter 'head' phantom or a 32 cm in diameter 'body' phantom. For defining absorbed radiation for Cardiac CT, a body phantom with imbedded radiation measuring 'probes' is used as illustrated in Figure 3. The scanner is set up to simulate a patient scan, but the phantom is the target. The 'absorbed dose' defines the average energy absorbed per meter of surface area and is expressed in the standardized unit of mGy (milli-Grays). Using the

CTDI, the weighted average of radiation to the probes defines or estimates the absorbed radiation dose. Changes in the mA, kV, and 'pitch' for simulated patient scanning must be accounted for in defining the CTDI/absorbed dose. As noted in Part I, the 'pitch' for general body scanning is 1.0 or higher, but for cardiac CT, the pitch is generally at 0.5 and the overlapping of x-rays increased the radiation exposure. Additionally, the phantom scans should be done at realistic patient simulated values for mA and kV.

The next step is to then calculate the individual 'organ' doses (H_T), which is based upon the 'absorbed dose' measurement (from the CTDI) and a rather nebulous term called 'biological effectiveness'. This term on 'effectiveness' can be derived from a listing or 'look up table' and is used to account for different sensitivities of various organs to radiation exposure. The ICRP (International Committee on Radiation Protection) is the organization that provides data from such look up tables. For instance the 'biological effectiveness' value for bone marrow (which is moderately radio-sensitive) would be higher than the heart (which is relatively radio-insensitive). Biological effectiveness for breast tissue is considered to be about twice that of other surrounding solid organs or skin. Thus: H_T = [CTDI x 'biological effectiveness'] and is calculated for each organ exposed to incident radiation. This radiation dose is also listed or defined in terms of mGy.

The final determination is the 'effective dose'. This is essentially the dose to the patient and is considered to be the yard stick for radiation risk. However, one more parameter must be defined and this is the 'organ weighting factor' or W_T. Although a particular organ may be more radio-sensitive (for instance the bone marrow), the actual amount of the organ exposed would be important to define 'risk'. For instance in the chest, the amount of bone marrow in the sternum and other surrounding bones is only a portion of the bone marrow found in the entire body while the breasts are almost surely completely contained within the imaging field. The organ weighting factors, also available from the IRCP, relate to organ sizes in a general sense contained in the areas x-rayed.

The 'effective' dose to a given organ is then [W_T x H_T]. By summing the effective doses to all radiated organs, we then derive the 'patient effective dose' for the x-ray procedure involved as the sum of the organ doses: i.e., 'patient effective dose' = Σ (W_T x H_T).

As noted the absorbed radiation dose and the organ radiation dose are given in units of mGy; this is a measure of the physical aspects of radiation. However, the ICRP has designated the measure of the 'effective' radiation dose using the mSv (milli-Sievert). This unit is named after the Swedish medical physicist, Rolf Sievert, who did pioneering work on radiation dosage and the biological effects of radiation. Thus the physical measures of mGy and mSv are identical, but the designation of mSv represents the unit used as the yard stick for defining patient risk.

Of course, medical diagnostic imaging using x-rays of any kind will result in ionizing radiation exposure to the individual. Ionizing radiation refers to the highly-energetic particles or waves that can detach (ionize) at least one electron from an atom or molecule. Other more ubiquitous sources of radiation, and thus contributing to 'risk' in a given individual for their lifetime, come from cosmic radiation (your personal exposure varies throughout the world and is based largely on geomagnetic field strength, altitude, and solar cycles). You get additional cosmic radiation every time you fly on an airplane. Some data suggest that airline pilots receive more average radiation exposure than any other worker, including those working in nuclear power plants. Of course, direct solar radiation (from the sun) is variable also depending on where you live and is a well known major concern in the development of skin cancer. There are also terrestrial sources of ionizing radiation from radioactive atoms such as potassium, uranium, and thorium. Radon gas is produced from the decay of radium and seeps into the soil and may be the largest single source of radiation dose to any living person and has been suggested to be the second largest cause of lung cancer in America, after smoking. Human-made resources of ionizing radiation (in addition to medical imaging) include tobacco, ophthalmic glass, television and computer monitors, luminous watches, airport security systems, smoke detectors, road construction materials, electron tubes, fluorescent lamp starters, and gas lantern mantles. A very interesting web site can give you an idea of your own personal radiation exposure; look up: http://www.ans.org/pi/resources/dosechart/.

Below is a table of common sources and effective amounts of ionizing radiation (mSv) from diagnostic imaging and daily living.

Source	**mSv**	**Source**	**mSv**
Cardiac CTA (maximum)	10-14	Mammogram (women)	0.7
Heartscan (calcium scan)	1-2	Airplane Trip to Asia	2
Thallium Stress Test	14-25	'Background'/year (USA)	2.5-3.5
Diagnostic Cardiac Catheterization	5-7	Barium Enema X-ray	10
MIBI Stress Test	8-12	Ru^{82} PET Scan	15

Of course the biggest source of mass radiation exposure has been the explosion of nuclear weapons on Hiroshima and Nagasaki and of course other nuclear accidents such as Chernobyl. In fact the consequences to the exposed populace surrounding these massive radiation sources has been the major impetus to provisions of radiation safety as discussed in Part III to follow.

Part III: Principles of ALARA and Radiation Safety in Cardiac CT

In Parts I and II we discussed how MDCT works and how to estimate effective radiation dose (to the patient) using Cardiac CT. In this Part III we will introduce additional terms related to radiation safety in medical imaging and for Cardiac CT in particular.

In the 1970's researchers noted an increased frequency of solid tumors in survivors of the atomic bombs dropped on Japan in 1945. It had been known for some time that the acute and high doses of radiation from the bombs AND the increased background radiation around Hiroshima and Nagasaki were associated with the development leukemia. The bone marrow is especially sensitive to whole body radiation and it had been known for nearly a century that high and prolonged radiation exposure could increase risk for lymphoma as well. It is been suggested that Marie Curie (the discoverer of radium) and perhaps her daughter both died of radiation induced leukemia. However, the discovery of 'solid' tumors (e.g. lung, breast, reproductive system) after radiation exposure was a surprise.

Data demonstrated that the incidence of solid tumors in survivors of atomic bomb attacks was as much as 50% greater than that of the non-exposed population. It has been estimated that those with cancers had > 100 mSv acute exposure; however, nobody has been able to absolutely demonstrate whether similar results would occur with smaller doses.

Committees including the ICRP (International Commission on Radiological Protection) were formed and soon suggested that *"As any* [ionizing radiation] *exposure may involve some degree of risk recommends that all unnecessary exposures be kept as low as is reasonable achievable…"* This ALARA is the basis for radiation protection.

ALARA (as low as reasonably achievable) is based on the generally accepted notion that higher doses of radiation are linked to both short-term and long-term effects on the human body and that one can extend this concern, using the LNT (linear, non-threshold) hypothesis stating: 'any radiation, no matter how low, carries with it a certain level of risk proportional to exposure'. In contrast studies done in airline pilots and health care workers exposed to frequent doses of radiation have not shown clear evidence for increased cancer rates. But, because of the relatively long time lines between acute exposure and development of solid cancers (as noted, at least 12 years in survivors of Hiroshima), concerns have been raised, especially in younger individuals who would have longer life spans after radiation exposure. As part of this concern another term came into being – LAR – which is 'lifetime attributable' risk.

There is no clear agreement that the risks of radiation exposure using diagnostic medical imaging significantly increases an individual's LAR, but there are data extrapolated from nuclear explosions and 'accidents' that have been scaled down to the amounts of exposure that might come from a Cardiac CTA. Although there is much discussion on the potential 'accumulative' risk of repeated diagnostic medical examinations – there are no scientific data to deny or refute the inferred increased risk. The acute exposure to 100 mSv of gamma radiation from a nuclear explosion as well as the continued radiation exposure from the decay of the isotope versus the acute exposure to 2-15 mSv of gamma radiation from a diagnostic Cardiac CT may be different, but may also have similar dose related consequences in accord with the LNT hypothesis.

A recent scientific analysis was published regarding the LAR for solid tumor cancers of men and women exposed to a single Cardiac CT examination (JAMA 2007;298;317-323). As with other investigations, the data were extrapolated using the LNT hypothesis. The effective radiation dose for a retrospective, ECG-gated Cardiac CT in women (14 mSv) is higher than men (10 mSv) due largely to the amount of breast tissue present and the known higher radio-sensitivity of the breast (recall how we determined effective dose related to the 'biological effect', H_T, and the 'organ weighting factor, W_T, as discussed in Part II). The report indicated the LAR is highest in younger individuals who would have more time to develop the consequences of ionizing radiation (as noted previously, solid tumors were found in atomic bomb survivors at a 12 or more years after acute exposure) and declines rapidly as initial exposure age increases. The theoretical results indicated small but not negligible risks. For example the LAR for a 20 year old woman receiving her first Cardiac CTA is 0.7% (this translates into one additional cancer in 143 Cardiac CT scans) while for a 20 year old man the LAR is 0.19% (1 in 526). At age 40 the LAR in women would be 0.35% (1 in 285) and in men 0.099% (1 in 1,010). The analysis also indicated that using ECG-dose modulation (see Part IV for details), the effective radiation dose in women and in men would be cut nearly in half (7.4 mSv in women and 5.4 mSv in men) and would also effectively cut the LAR for cancer by a similar proportion. The conclusions of this paper underscore and emphasize the principles of radiation safety, as noted below:

1. Justification of dose (is the incremental diagnostic value of performing Cardiac CTA versus the risk of the study sufficient AND are there alternative non-ionizing radiation methods that may provide the same or similar results)
2. Make radiation dose as low as reasonably achievable (ALARA)

One might ask – why the fuss here? The use of Cardiac CTA is largely to diagnose coronary artery disease, which is much more common in a 50 year old than a 30 year old – suggesting that such a test would be uncommonly performed in a younger person (with higher LAR) than in an older person (with lower LAR). But the issue goes beyond just Cardiac CT and extends to the rapid adoption of CT in general for diagnostic imaging in all age groups, including pediatrics. Thus, the real issue is not just Cardiac CTA, but CT in general and the potential increased LAR as the utilization of these moderately high radiation dose examinations increases. This is a fundamental question that is being addressed by major regulatory agencies. In 1980, the number of CT examinations was estimated at 3 million per year; but in 2008, it is estimated that this will be 68 million/year. To give you a better feeling for the rapid increase in the use of

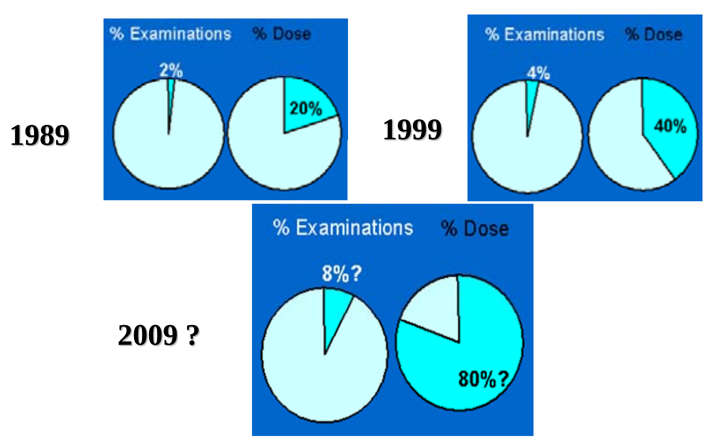


Figure 4

CT scanning, consider Figure 4. In 1989, CT represented only 2% of all radiological procedures, but accounted for 20% of the total burden of medical radiation to the populace. In 1999, CT represented 4% of the diagnostic radiology procedures and accounted for 40% of the total medical radiation. It is estimated that in 2009 these numbers will increase to 8% of the total exams and account for up to 80% of the diagnostic imaging radiation dose to patients.

In Part IV we will discuss various methods that can help reduce the radiation dose to the patient and still maintain the diagnostic power of Cardiac CT.

Part IV: Methods to Reduce Effective Radiation Dose from Cardiac CT

In Part I, we discussed how CT scanners and in particular Cardiac CT scanning works. In Part II we discussed how to measure radiation dose using Cardiac CT; and in Part III we discussed the issues of radiation effects on the body and principles radiation safety. In this final, Part IV, we discuss various scanning/imaging protocols using Cardiac CT that will maintain diagnostic accuracy, but lower effective radiation dose to the patient.

The methods available to reduce effective radiation dose to the patient while performing Cardiac CTA fall into five categories [Note, there is actually some overlap for these methods, as discussed below]:

1. reduce total exposure time/area
2. reduce radiation from the x-ray tube
3. reduce or eliminate spiral image overlap
4. external shielding
5. a combination of two or more of the above

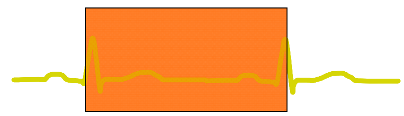
Reducing Total Exposure Time/Area:

There are currently 2 methods to reduce the total exposure time/area and involve prospective ECG gating and limiting the volume of the chest irradiated.

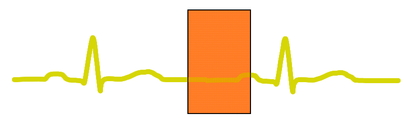
Traditional Cardiac CT, related to issues of continuous cardiac motion during each cycle and true temporal resolution of the individual scanner, has used ‘retrospective’ ECG gating with spiral (overlapping) imaging which also produces the ‘maximum’ effective radiation to the patient. That is, the images are acquired throughout many cardiac cycles and all data are available to the array processor (see Part I) for reconstruction. This then allows searching the images for the phase of the cardiac cycle in which there is the least motion thus facilitating diagnostic Cardiac CT coronary angiography. Using spiral retrospective gating, data throughout the cardiac cycle are also available to investigate the dynamics of cardiac global and regional function. However, x-ray exposure is continuous during the image acquisition protocol and thus effective radiation is the highest.

Several of the CT scanner manufacturers have introduced spiral ‘prospective’ ECG gating. This is actually a re-introduction of the standard method used for Electron Beam cardiac CT (EBT) first developed and validated in the 1980’s. But EBT had a true temporal resolution of actually 86 msec [also was a true ‘step and shoot’ scanner, see later] compared to current MDCT with approximately 160 msec temporal resolution (not including the newer ‘dual’ source scanner). When the heart rate is roughly 55-60 beats per minute, it has been shown that images from the diastasis phase of the cardiac cycle (roughly 65% to 75% of the actual RR-interval) are generally motion free and, by limiting the actual time that the x-ray source is ‘turned on’ to only a portion of the ECG cycle, would reduce the actual time of x-ray exposure to the patient. Studies have shown that prospective gating of the standard spiral Cardiac CT examination can reduce total effective radiation doses by 50%.

In prospective gating the scanner gantry (see Part I) continues in the usual spiral pattern, with a pitch generally at 0.5, but the x-ray source is 'gated' to be turned on only during diastasis. This is illustrated in Figure 5 as compared with retrospective gating.



"Cine" (movie) Spiral Mode - Retrospective



Retrospective Spiral

Figure 5

The advantage of performing retrospective gating using Cardiac CT is that multiple phases, at any chosen ECG phase, can be reconstructed and this can be used to also define global and regional cardiac function; using prospective ECG gating only data during diastasis are available for review. The diagnostic accuracy in defining detailed coronary anatomy, if done properly, is not different using either method, but cardiac function cannot be assessed using prospective gating. Of course if assessment of LV function is required, this could be done effectively using methods that are non-radioactive, such as two-dimensional echocardiography or MRA (magnetic resonance angiography).

However, there is one additional caveat here using prospective gating and that is proper heart rate control during image acquisition. In order to perform prospective gating in a diagnostic mode and a single source scanner it is imperative that the heart rate be as close to 60 beats per minute (or less) and this requires judicious use of beta-blockade in most individuals. Even at a heart rate of 65 beats per minute, it is not uncommon for the most motion free portion of the cardiac cycle to be best during late or end systole (35% to 50% of the cardiac cycle). Thus in the attempt to limit the effective radiation dose to the patient using prospective gating, the radiation dose might be 'wasted' (which in our opinion is a worse scenario) since the examination may not be diagnostic using the only images available. In our experience, it is not uncommon for the 'best' phase to interpret the study to be elsewhere than during diastasis. Utilizing retrospective gating allows one to have available the dataset for another phase that might allow the study to be diagnostic, even if the heart rate increases during the study to 70-75 beats per minute. Proper use of beta blockers during Cardiac CT is, in a real sense, a hidden method to reduce effective radiation dose to the patient. The above discussion is for standard single x-ray source scanners and does not absolutely apply to the use of dual source Cardiac CT (with improved temporal resolution) which allows diagnostic scanning of individuals using prospective gating even at heart rates up to 75 beats per minute.

The second method to reduce radiation to the patient is to limit the areas (size) of the radiation field (i.e. the total number of thin sectioned slices). The total number of slices required to interpret a retrospective or prospective gated Cardiac CT are the same for examining native coronary arteries; however, in the case of defining bypass graft anatomy or other areas of the ascending/thoracic aorta and the pulmonary arteries (for identification of pulmonary emboli) the number of slices must be increased to expand the field of view. In many centers that routinely perform Cardiac CTA to assess bypass grafts, scanning begins at the bracheocephalic trunk in order to identify the origins of the internal mammary arteries. In our experience (and opinion) such extended imaging is not necessary since the causes for occlusion of the right or left internal mammary arteries are rarely at their origins. Just the same, imaging of bypass grafts does require at least extending the field to the aortic arch; while imaging of the native coronary arteries only involves extending the imaging to the proximal ascending aorta, roughly at the level of the pulmonary artery bifurcation.

Reducing Radiation Dose from the X-ray Tube

As discussed in Part II of this paper, effective radiation doses are very much dependent on x-ray tube current (mA) and x-ray tube voltage (kV). Reducing mA or kV, or both, during Cardiac CT can substantially reduce the effective radiation.

Standard Cardiac CT is done most often using a setting at 120 kV. Studies have shown that by reducing this to 100 kV, the effective radiation dose (using retrospective or prospective gating) can be reduced by as much as 30% without sacrificing diagnostic accuracy. The caveat here is that this reduction in kV works well in non-obese patients, say <80-85 kg in weight. In obese patients such a method would more likely result in 'wasting' radiation as the images may be too noisy as more penetrating power is needed (this would be analogous to taking a photograph in dim light without the use of a flash).

One additional method, available on all 64-slice scanners, is called ‘ECG dose modulation’ or ‘ECG dose pulsing’. This involves reducing the mA during certain phases of the cardiac cycle during retrospective gating. If the beta-blockade is adequate, dose modulation can be employed during systole and returned to full power during diastole. The systolic images are of course, less optimal, but cardiac function can still be assessed and, depending on the reduction in mA, these images are available for high resolution images of the cardiac cycle. It is not uncommon for a patient to have a heart rate of 55-60 beats per minute during scan set up only to have the heart rate increase dramatically when they get the initial ‘warm’ feeling from administration of the iodinated contrast. ECG dose modulation can decrease the effective dose to the patient from 30%-50% (see comments on LAR reduction using dose modulation in Part III). This protocol applies of course only to retrospective gated Cardiac CTA and has no application to prospective gating.

Reducing or Eliminating Spiral Scan Overlap

A major source of effective radiation dose is the spiral overlap which is a function of the pitch and the amount of physical coverage for spiral of the scan. The total area to be imaged for the native coronary arteries is about 12 cm. Using a 64-slice spiral scanner at a pitch of 0.5, requires about 8-10 spirals of the gantry. If one were to, in some manner, eliminate or reduce the overlapping spirals, radiation dose can be reduced.

Several of the CT scanner manufacturers have introduced the ‘step and shoot’ method. Using this method the patient couch does not increment until each spiral has been completed and then the couch increments and waits for the protocol designated next cardiac cycle (or portion of the cardiac cycle for prospective gating) and then repeats the scanning. Variations in heart rate during the scanning sequence can be quite detrimental to the subsequent tomographic interpretations as this would require putting together the scans in a manner of a jig saw puzzle. Absolute control of the heart rate and limited variability (again, the judicious use of beta-blockers) and a steady heart rate is crucial; however successful application of this process has been shown to reduce the effective radiation doses by 50% or greater. This is most commonly applied using prospective gating for perhaps up to 70% reduction in effective radiation dose compared to standard retrospective spiral gating.

Several manufactures have begun to introduce MDCT scanners with the capability of taking 256 or 320 slices per gantry rotation. The spatial resolution and the temporal resolution are the same or only slightly better than that of 64-slice scanners, but the elimination of the need for spiral overlap can greatly reduce the effective radiation. If this method is combined with prospective gating, some have noted that diagnostic Cardiac CT can be performed with effective radiation doses of 2-3 mSv.

External Shielding

A final method to reduce incident radiation is to use an external shield. The most common concern for the potential LAR (see Part III) is to limit the radiation dose to the breast in women. Several types of shields are available and the manufacturers suggest that breast radiation can be substantially reduced with their device (claims of 25%-70% reduction). Figure 6 shows the application of an external, foldable ‘breast shield’ during a CTA done in a young woman. In this example there is some ‘noise’ in the images of the chest wall, but the coronary scan is not affected and is diagnostic.

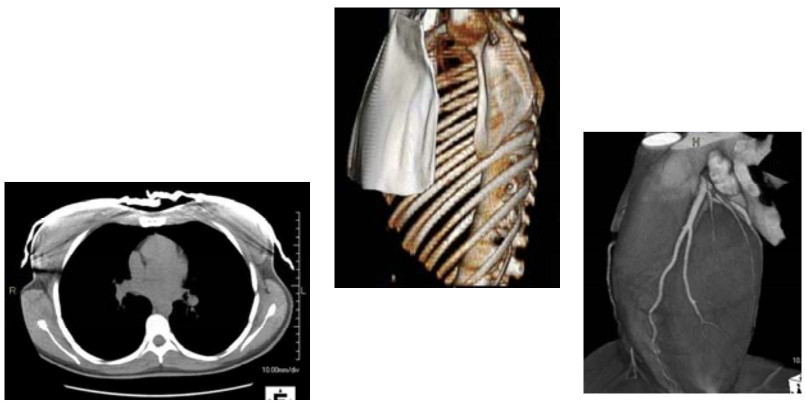


Figure 6

In this compendium, divided into 4 parts, we have discussed a number of principles of CT scanner operation and radiation safety which are essential to master in order to be ACCF and ACR Certified at Level II and Level III in Cardiac CTA. In the future additional topics will be presented in this continuing educational series in successful application of Cardiac CT in clinical practice.