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Radiation Safety: Radiation Dosimetry and CT Dose Reduction Techniques

Kai H. Lee

Introduction

The ability of modern multidetector CT scanners with submillimeter resolution, subsecond rotation time, and large volume imaging has resulted in widespread utilization of cardiovascular computed tomographic angiography (CCTA) [1]. However, the widespread use of CCTA has also raised concerns about the radiation dose to the patients. The National Council on Radiation Protection, NCRP Report No. 160, reported that the radiation exposure to the United States population due to medical sources increased more than 7 times in the 20 years between 1986 and 2006 [2]. Although in 2006 CT constituted about 10% of the diagnostic examinations that utilize X-rays, it contributed to nearly 50% of the population dose [2–4]. Based on our current knowledge of radiation biology, the deleterious effects of radiation are cumulative and medical radiation is increasingly a significant contributor to the amount of radiation accumulated in a person's lifetime [1, 2, 5]. The risk of cancer from radiation exposure is especially worrisome in children and young women who received multiple CT examinations early in their lives. For example, studies found that one CT examination of the female chest gives as much radiation as 10 mammograms to each breast [6]. Therefore, the practitioners of CT must be constantly aware of the risks of radiation and strive toward applying the lowest dose to the patient consistent with the clinical study.

One of the difficulties confronting physicians when evaluating the radiation safety of a CT procedure is the plethora of terms used to quantify the amount of radiation given to the patient. Thus, this chapter has two aims. The first aim is to explain the fundamental concepts of radiation dosimetry relevant to cardiac CT examinations. The second aim is to describe the techniques that physicians may exercise to control radiation dose to their patients.

Fundamentals of Radiation Dosimetry

Absorbed Dose

The International System unit (SI units) of radiation dose measurement is the *Gray* (Gy) [7]. However, Gray is a large unit of radiation. When evaluating radiation dose from CT examinations, two smaller divisions of the Gray are commonly used. The two subunits of Gray are the milliGray (mGy) and the centiGray (cGy):

$$1\,\text{mGy} = 10^{-3}\,\text{Gy}$$
$$1\,\text{cGy} = 10^{-2}\,\text{Gy}.$$

In the United States, the SI units are quoted in the literature, but the traditional unit *rad* is utilized in routine radiation safety practice [7]. The unit *rad* is an acronym for *r*adiation *a*bsorbed *d*ose. The conversion between the traditional and the international systems of radiation dose measurements is simple: 1 Gy = 100 rad. It follows that 1 cGy = 1 rad.

Equivalent Dose

The severity of biological damage depends not only on the amount of radiation absorbed. It also depends on the type of radiation absorbed. For example, 1 Gy of neutron radiation is 10 times more damaging than 1 Gy of X-rays [8]. A unit of measurement is required that takes into account the effectiveness of different types of the radiation in producing biological damages. The biological equivalent unit used in radiation protection is the Sievert (Sv), and the traditional unit is the rem (acronym for radiation equivalent man). In SI units, the biological equivalent dose H is equal

to the radiation absorbed dose D measured in Gy multiplied by the radiation damage weighting factor W_r, i.e.,:

$$H(\text{Sv}) = D(\text{Gy}) \times W_r,$$

when using the traditionally units, the biological equivalent dose in rem is equal to the absorbed dose in rad multiplied by the quality factor Q, i.e.,:

$$H(\text{rem}) = D(\text{rad}) \times Q,$$

where the quality factor Q serves the same function as W_r to account for the relative effectiveness of different types of radiation in producing biological damage. The numerical values of Q and W_r are in fact identical for the same type of radiation. Some typical weight factors are given in Table 3.1.

The values of W_r or Q are proportional to the density of ionization created by the incident radiation along its path of travel in tissue. For X-rays, gamma rays, beta particles, and electrons from radioactive materials, the density of ionization created in tissue is relatively low. The weighting factor W_r equals to 1. Thus, when working with X-rays from CT, the equivalent dose and absorbed dose are numerically equal, i.e., 1 Sv = 1 Gy, and 1 rem = 1 rad. For neutrons, the weighting factor $W_r = 10$. The equivalent dose for 1 Gy of neutron absorbed dose equals to:

$$H(\text{Sv}) = 1(\text{Gy}) \times 10 = 10\,\text{Sv}.$$

This example shows that neutrons are 10 times more damaging to the human body than X-rays for the same absorbed dose. In other words, 1 Gy of neutron produces 10 times greater risk than 1 Gy of X-ray.

For radiation protection purposes, subunits of Sievert and rem are used. Subunits of Sievert and rem are the milliSievert (mSv), and millirem (mrem), respectively. Since both the traditional and SI units are used in the US, it should be noted that 1 mSv = 100 mrem. A convenient method to convert an equivalent dose given in SI unit to the traditional unit is to multiply the SI values by 100 and change the unit name from Sv to rem but leave the numeric prefix unchanged.

In summary, *absorbed dose* is the quantity of radiation energy deposited per unit volume of tissue. The equivalent *dose* is a measure of biological damage equal to the absorbed dose modified by a weighting factor according the relative effectiveness of the incident radiation to produce biological damage.

Table 3.1. Radiation weighting factors

Type of radiation	Weighting factor
X- and gamma rays, electrons, positrons	1
Neutrons	10
Protons	2
Alpha particles	20

Table 3.2. Typical annual whole body radiation dose

	mSv	mrem
Nuclear medicine technologists	1.2	120
Airline flight crews	2.2	220
Nuclear power plant workers	5.5	550
Interventional radiologists	18	1,800
Interventional cardiologists	16	1,600
Natural background radiation	2.5	250
Regulatory limit on the occupational workers	50	5,000
Regulatory limit on the general public	1	100

Given this metric to quantify radiation, Table 3.2 lists the average equivalent dose received annually to the total body by workers in the various occupations [9–12]. The table also gives the natural background radiation and the regulatory limits on radiation exposure as reference to occupational exposures. It is interesting to note that airline flight crews, who are not classified as occupational radiation workers, receive annual equivalent dose from the cosmic rays of nearly twice as much as the nuclear medicine technologists who routinely handle radioactive materials on the job.

Effective Dose

Unlike the absorbed dose and the equivalent dose, the effective dose is not a physically measurable quantity. The effective dose is an imaginary total body dose. It is calculated from the absorbed dose given to any part or parts of the body. That is, the effective dose is the equivalent whole body dose if the radiation given to a partial body exposure such as a CCTA study was spread uniformly across the entire body. The purpose of the effective dose is to translate a partial body exposure such as a cardiac CT scan to an equivalent uniform total body dose to assess the risk of carcinogenesis and genetic defects.

It is important to calculate the effective dose from a given medical procedure, such as CCTA, because our current knowledge of the risk of radiation-induced carcinogenesis and genetic defects is based on data collected from the total body exposed uniformly to certain doses of radiation. In order to estimate the risk resulting from CT of the chest for example, one must translate the partial body irradiation to an equivalent whole body exposure in order to utilize the database for risk estimates. To do so, a mathematical model is used to compute the doses to all other organs resulting from radiation scattered from CT of the chest. These computed organ doses are then multiplied by a risk factor according to the sensitivity of each organ to radiation. Summation of the product of these computed organ doses and their associated risk weighting factor is called the effective dose. That is, the effective dose E is computed using the equation:

$$E = \Sigma H_i w_i,$$

where E is the effective dose, H_i is the dose equivalent to a given organ, and w_i is the risk weighting factor for that organ.

The effective dose is thus a weighted sum of the computed doses to all organs in the body. A table of weighting factors for different body organs is given in a report by the International Commission on Radiation Protection [8, 13].

One may interpret the effective dose as a calculated equivalent dose of radiation given to the entire body that would be required to produce the same risk of cancer and genetic damages as a dose of radiation delivered to a localized region of the body as in a CT examination. In other words, the risk from a part of the body exposed to a given dose of radiation is the same as the total body uniformly receiving the effective dose. Thus, the effective dose is an extrapolated whole body dose from a partial body dose. As such, the effective dose is a computed value rather than a physically measurable quantity. The effective dose is calculated to serve as a common denominator for comparison of stochastic risk between different medical or nonmedical exposures to radiation.

With an understanding of the concept of the absorbed dose, equivalent dose, and effective dose, the next step is to learn how to calculate the effective dose from CT examinations.

CT Dosimetry

Special dosimetric techniques have to be developed for measuring CT doses because the geometry of the X-ray field employed in CT scans is very different from the conditions of conventional radiographic exposures [14, 15]. The fundamental parameter developed for CT dosimetry is the Computed Tomography Dose Index (CTDI). From the CTDI, the dose-length product (DLP) is calculated and then used to derive the effective dose (E) for risk comparison.

CTDI

The CTDI, or specifically the $CTDI_{100}$, is measured using a dosimeter of 100 mm length in a cylindrical acrylic phantom of 16- or 32-cm diameter to simulate a head or body (Figure 3.1). Four measurements are made in the periphery and one in the center of the phantom. The four peripheral measurements and the central measurement are used to compute the weighted average of the $CTDI_{100}$ in the phantom as follows:

$$CTDI_W = 0.87 \times [2/3 CTDI_{100}(\text{periphery}) + 1/3\, CTDI_{100}(\text{center}),$$

where $CTDI_w$ is the weighted average of $CTDI_{100}$ in the phantom.

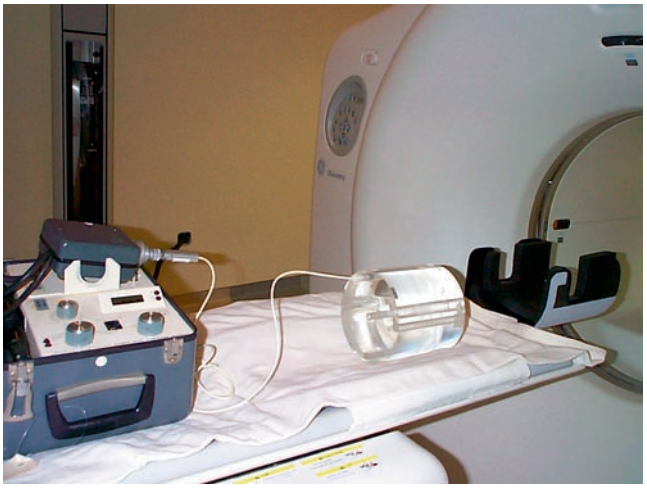

Figure 3.1. A 100-mm length cylindrical acrylic phantom used for measuring the $CTDI_{100}$.

The factor 0.87 is a conversion factor to relate dose measured in air to dose in soft tissues. The $CTDI_w$ can be interpreted as the average absorbed dose in the cross section of the patient from one axial scan.

For helical scans, the average dose may be greater than or less than the dose from an axial scan. If in a helical scan the table moves the patient slowly through the gantry, the average absorbed dose in a transverse section of the patient may be greater than that in an axial scan due to the X-ray beam overlapping on the patient in successive rotations of the X-ray tube. Conversely, if the table moves at a high speed, the X-ray beam passes through the patient in a nonoverlapping helical path as shown in Figure 3.2. There would be less radiation delivered to the scan volume because of the gaps between tracks of the X-ray beam. That is, for CT scans done with a pitch of less than 1, there is overlapping of the X-ray beam as it passes through the patient, and the patient receives a higher dose. For scans done with a pitch greater than 1, the X-ray beam does not overlap, and the patient receives less radiation. Of course, there is greater image noise for scans done with pitch greater than 1. The clinicians will have to make a compromise between image noise and the patient dose.

The $CTDI_{vol}$ was therefore developed to compute the average absorbed dose in the scan volume taking into account the variable overlaps in the spiral path of the X-ray beam. The $CTDI_{vol}$ is calculated using the following equation:

$$CTDI_{vol} = CTDI_W / \text{pitch}.$$

Pitch is a dimensionless unit equal to the distance the table traveled during one complete rotation of the X-ray tube divided by the width of the X-ray beam at the axis of rotation. For a multislice CT, the pitch is defined as follows:

$$\text{Pitch} = D / nT,$$

where D is the distance the patient table moved in one rotation of the X-ray tube, n is the number of slices produced in

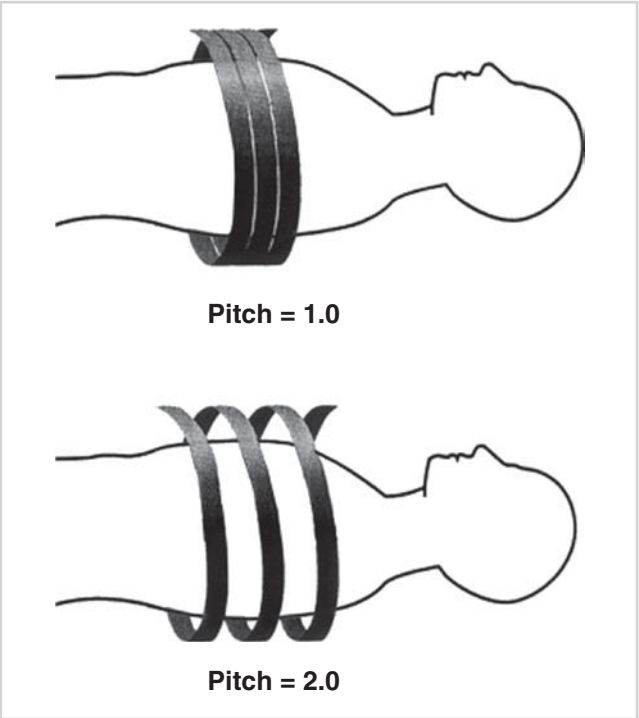

Figure 3.2. Schematic demonstrating the concept of pitch. With a pitch of 1, there is no overlap or gap. With a pitch of greater than 1, the X-ray beam passes through the patient in a nonoverlapping helical path with less radiation delivered to the scan volume, but with greater image noise. With a pitch of less than 1, there is overlap in the X-ray beam path, and the patient receives a higher radiation dose, but with less image noise.

one tube rotation, T is the thickness of each slice measured at the axis of rotation, and the product nT represents the width of the X-ray beam that sweeps around the patient during a CT scan.

Dose-Length Product

One final measurable dosimetric parameter for CT is the DLP. The DLP is proportional to the total amount of X-ray energy deposit in the scan volume. DLP is defined as:

$$\mathrm{DLP} = \mathrm{CTDI}_{\mathrm{vol}} \times \mathrm{scan\,length}$$

The cross-sectional area of the patient is implicitly included in $\mathrm{CTDI}_{\mathrm{vol}}$ by multiplying the $\mathrm{CTDI}_{\mathrm{vol}}$ by the scan length. DLP thus describes the volume of tissues irradiated and the total amount of X-ray energy deposited in that volume. DLP is an important risk indicator because the severity of biological damage from radiation depends not only on the quantity and type of radiation given, but on the volume of tissue irradiated as well. For example, a radiation oncologist could deliver 7,000 cGy of X-rays to a small region surrounding the prostate gland to cure a patient with prostate cancer. If 7,000 cGy was delivered over the entire body, the person would most certainly die from the absorbed dose. Thus, the risk of radiation increases with the volume of tissue irradiated. The two descriptors of CT dosimetry, $\mathrm{CTDI}_{\mathrm{vol}}$ and DLP, serve to quantify the amount of X-ray energy absorbed per unit mass of tissue, the volume of tissue exposed to radiation, and thus the total amount of energy deposited in the scan volume. Because $\mathrm{CTDI}_{\mathrm{vol}}$ and DLP are such important indicators of the risk from a CT procedure, the values of $\mathrm{CTDI}_{\mathrm{vol}}$ and DLP have been displayed on the control console in all CT systems manufactured since year 2000.

The Effective Dose from CT

For other imaging procedures, the effective dose has to be computed from an elaborate computer model. The effective dose from CT, however, can be calculated from the DLP using a simple formula developed by the ICRP [16, 17]. The effective dose is conveniently calculated by multiplying the DLP by the corresponding conversion factor shown in Table 3.3, i.e.,:

$$\mathrm{E(mSv)} = \mathrm{DLP} \times \mathrm{CF}$$

where CF is the conversion factor for the corresponding CT procedure.

By using the $\mathrm{CTDI}_{\mathrm{vol}}$, DLP, and conversion factors in Table 3.3, some typical values of the effective dose from different CT procedures are shown in Table 3.4. The effective doses from other radiologic examinations are also shown in the table for comparison [18, 19].

CT Dose Reduction Techniques

The effective doses in Table 3.4 show that CT is a high dose procedure compared with other X-ray examinations. As mentioned, CT examinations consisted of about 10% of all diagnostic examinations that utilize X-rays, but contributed to nearly 50% of the medical radiation to the population. The contribution of CT dose to the population is expected to continue to rise given the expanding use of CT. It is therefore imperative for the practitioners of CT to

Table 3.3. CT effective dose conversion factors

Region of body	Conversion factor
Head	0.0023
Neck	0.0054
Chest	0.0170
Abdomen	0.0150
Pelvis	0.0190

Table 3.4. Typical effective doses of radiological examinations

Radiologic examination	Typical effective dose (mSv)
Head/neck CT	2–5
Chest CT	5–7
Abdomen/pelvis CT	8–11
Coronary CT angiogram	5–15
Coronary bi-plane angiogram	3–10
PET cardiac viability per 370 MBq	4–7
MIBI cardiac stress/rest per 1.3 GBq	10.6
PA chest X-ray	0.02
Skull X-ray	0.07
Lumbar spine	1.3
I.V. urogram	2.5
Upper GI	3.0
Barium enema	7.0

obtain the desired diagnostic information using low dose techniques whenever possible.

Several recent studies reported that radiation dose in cardiac CT can be reduced by more than 50% simply by modifying the basic technical factors in their CT scanning protocols without having to invest in new equipment [20–24]. The technical factors available to nearly all CT scanners for dose reduction include limiting the scan length, lowering the tube voltage, modulating the X-ray tube current, minimizing scan time, and using prospective ECG gating.

Limiting the Scan Length

The effective dose to a patient is proportional to the dose-length-product DLP, which in turn is proportional to the craniocaudal length being scanned. One study reported the use of the calcium score protocol as a guide to determine the minimum scan length for each individual patient in a CCTA examination [25]. Prior to CCTA, the patient was given a prescan using the calcium score protocol. The start of the scan was then determined from the calcium score images at 1 cm above the visualized top of the coronary arteries and stopped at 1 cm below the posterior descending artery. The dose savings from limiting the scan length more than offset the dose from the calcium score scan. The effective dose was reduced by as much as 30%.

Reducing the Tube Voltage

Coronary CT angiography protocols commonly use a tube voltage of 120 kV for all patients. Reducing the tube voltage from 120 to 100 kV for patients weighing 185 pounds or less has been reported to reduce the effective dose by 30–40% to the range 5–12 mSv from 9 to 17 mSv [20, 21, 24]. Other studies showed that, for thin patients, the tube voltage can be further reduced to 80 kV without degrading the diagnostic accuracy, and the dose savings was as high as 80% [26]. Although the tube voltage affects the image contrast, the above studies showed that the overall image quality was preserved when matching the size of the patient with the reduced tube voltage.

Modulating the Tube Current and Time Product

Radiation dose to the patient is approximately directly proportional to the product of the tube current measured in mA and the tube rotation time in seconds. While options for varying the tube rotation time are limited for CCTA, there is great latitude in the choice of tube current. Similar to reducing the tube voltage, there is a delicate balance between the desire to lower the mA for dose reduction, and the need to apply sufficient mA to keep the image noise from degrading the image quality. When the mA is reduced, the intensity of the X-ray beam passing through the patient is decreased, therefore reducing the patient dose. But by reducing the mA, the image noise is increased and the overall quality of the images is reduced because fewer photons, less data, are available to reconstruct the images.

The question becomes one of deciding the optimum mA to minimize dose to the patient while still producing images of acceptable quality for diagnosis. CT manufacturers have responded to this question by incorporating dose reduction options into their system. The various dose reduction techniques available in modern CT scanners are actually different implementations of automatic exposure control (AEC) that have been in use for decades in fluoroscopy and radiography. The AEC controls dose to the patient by modulating the X-ray beam intensity according to the patient's anatomy to produce the desired image quality. The intensity of the X-ray beam emitted from the X-ray tube and subsequently sent towards the patient is directly proportional to the mA across the tube. By modulating the electron current mA according to the thickness of tissues that the X-ray beam must pass through, the desired image quality can be maintained without imparting unnecessary radiation to the patient. For example, the CT scanner can automatically raise the mA to produce a more intense X-ray beam to pass through the abdomen, and reduce the mA to produce a less intense X-ray beam when passing through the lungs.

There are three conditions under which the AEC can modulate the tube current (mA) to produce the desire image quality, and in the process reduce the unnecessary radiation to the patient. First, the AEC can be programmed to adjust the mA along the long axis of the patient so that the mA is reduced when the X-ray beam passes through large volume of air in the thorax, and is raised when the beam goes through the more attenuating soft tissues in the abdomen and pelvis. Second, the mA can be adjusted in the transverse plane of the patient according to the tube angle during its rotation around the patient. That is, the mA is reduced when the X-ray beam passes through the patient in the thinner AP and PA directions, and increased in the thicker lateral directions. Third, the overall tube

current can be adjusted according to the patient's size such that a lower mA would be used on pediatric patients and thinner adult patients.

There are three general methods used by the manufacturers in their dose reduction options [18, 27, 28], each with their own advantages and disadvantages. Implementation of the dose reduction option changes frequently with changes in the technology, but the principles behind the methodology remain essentially the same. The three common algorithms employed by manufacturers of CT scanners are AEC guided by image noise, AEC guided by reference image, and AEC guided by reference mAs.

AEC Guided by Image Noise

Image noise can be described qualitatively as the graininess of the image. Low noise images appear smooth with continuous shades of gray from the darkest to the lightest portions of the image. High noise images show characteristic salt and pepper grains interspersed throughout the image. Resolution of low contrast objects can be greatly impaired by image noise. Image noise is influenced by a number of factors, but is ultimately determined by the number of X-ray photons that contribute to the image. A simple way to quantify noise in an image is to calculate the percentage standard deviation. The standard deviation of an image is the square root of the total number of counts or dots that make up the image. The resulting standard deviation is then divided by the total counts in the image to arrive at the percentage standard deviation. The standard deviation of the number of counts in an image is an expression of the fraction of noise in an image. The percentage standard deviation as computed by the ratio of the standard deviation to the total number of counts in the image indicates what fraction of the image is occupied by noise. Low contrast objects are often obscured by the image noise.

The number of counts in a given CT image is directly proportional to the number of X-ray photons available for image formation, which in turn depends on the intensity of the X-ray beam striking the detectors and the length of time that the X-ray beam is on. The AEC modulates the beam intensity by varying the tube current. The exposure time is determined by the speed of the X-ray tube to make one rotation around the patient. The product of the tube current in mA and the rotation time in seconds is commonly called the mAs. The mAs selected for a CT scan determines the number of X-ray photons striking the patient. The higher the mAs, the greater the number of photons available to pass through the patient to reach the detectors, and therefore the lower the noise in the reconstructed images. When all the other technical factors are held constant, the image noise is inversely proportional to the square root of the mAs, i.e.,:

$$\text{Noise} \sim 1/\sqrt{\text{mAs}}.$$

During CT scans, the tube rotation time is fixed. The image noise guided AEC continually adjusts the mAs by varying the mA to maintain the same number of photons reaching the detectors, and hence keeping the image noise at a preselected level. For example, the AEC reduces the mA when the beam is passing through in the thinner anterior–posterior direction of the patient, and increases the mA when passing through laterally.

Patient dose optimization can be achieved by selecting a target image noise level appropriate for the particular CT procedure. The disadvantage is that the target noise level selected by the user may be lower than necessary for obtaining the diagnostic information, and result in giving unnecessary dose to the patient. A simple rule to remember is that reducing the mAs and hence the patient dose by a factor of 4 increases the image noise only by a factor of 2.

AEC Guided by Reference Image

This method of AEC is an extension of the constant image noise algorithm. Here, the target image noise is set on the AEC using a clinical image that is selected by the reader as of adequate quality for the given CT procedure. During scan, the mA at each tube position is adjusted by the AEC to yield an image noise approximating the noise in the reference image. The advantage of the reference image approach is that the target image noise is derived from a database of clinical images rather than some abstract percentage standard deviation. The major disadvantage of the reference image approach is the user inclination to select the most esthetic image as the reference image even though a less esthetic image will allow diagnosis. This results in the patient receiving a larger radiation dose than necessary.

AEC Guided by Reference mAs

This approach uses the mAs of a reference patient as a guide to modulate the mA for the actual patient. For a given CT procedure, a certain mAs that was found to produce images of acceptable quality on a reference patient is used as the standard of reference. From the attenuation profile measured on a scout view of the actual patient, the AEC adjusts the tube current at each tube position to compensate for the difference in attenuation between the actual and reference patient. The image noise is not maintained for different patient sizes. The technique relies on the experience of the user to select the proper level of tube current modulation for a given patient and CT procedure.

ECG Gating

Prospective ECG gating is based on the finding that least motion artifacts were found in the images reconstructed

from data in the ventricular diastolic phase. The X-ray beam is turned on only during middiastole to acquire the image data, and turned off during the other phases of the cardiac cycle. Prospective gating is also known as an ECG gated step-and-shoot technique. The most dramatic reduction of dose has been reported by using a prospective gating technique in combination with tube voltage reduction [24, 27, 29].

With prospective gating, the tube current is modulated by the ECG tracings to synchronize with the patient's cardiac cycle. Signals from the ECG monitor trigger the X-ray beam to turn on for data acquisition during diastole, and turn off when not acquiring data during the systolic and early diastolic phases. The table is then stepped to the next position for scanning in the predefined phase of the cardiac cycle. The heart is thus scanned in a sequential step-and-shoot fashion as shown in Figure 3.3.

Immediately before the scan, a sample of the ECG tracing is taken to measure the average R–R time interval. When the CT is set to scan, the AEC sets the clock to zero upon receiving an R-wave from the ECG monitor. The tube current is turned off until 70–75% of the cardiac cycle has elapsed. After this initial delay, the tube current is turned on to the maximum and maintains the maximum output for the next 10% of the cardiac cycle, corresponding to the time during which the heart is in the diastolic phase. The AEC subsequently turns off the tube current at the 80–85% mark that approximates the end of the diastolic phase to stop data acquisition. The time marker is reset to zero upon receiving the next R-wave. While the X-ray beam is off, the table moves the patient to the next scanning position. By activating the X-ray beam during only 10% of the cardiac cycle for data acquisition, the patient dose was reduced by 50–70% in comparison with retrospective gating that requires that the X-ray beam on continuously throughout the scan [22].

Although the advantage of prospective gating is dose reduction, images are only obtained at a certain percentage of the R–R interval. If there is motion artifact of the coronary arteries at this phase of the cardiac cycle, images may not be diagnostic. Retrospective gating can somewhat limit dose through dose modulation. With this technique, the tube current is ramped down during a certain portion of the cardiac cycle. Although the phases of the R–R interval during the phases of dose reduction may not be useful for coronary artery assessment, they may still be used for cardiac functional assessment. Both prospective gating and retrospective gating with dose modulation are problematic in patients with rapid or irregular heart rates.

Rapidly advances in MSCT technology have brought drastic changes to cardiovascular imaging techniques and lowering of radiation dose to the patient. Of particular interest to cardiologists is the introduction of 320-slice MSCT. Preliminary studies [27] using the 320-slice CT found that image data for the entire heart volume could be acquired in a single rotation, and obviates helical scans and bed indexing to acquire data over several cardiac cycles. When using a combination of kV reduction, prospective gating, and elimination of scan overlap of the X-ray path, dramatic reduction of the patient dose can be achieved without compromising the image quality.

Conclusion

High speed multislice CT with its high temporal and spatial resolution is increasingly applied for a wide spectrum of cardiac studies. Because of the high radiation dose associated with cardiac CT examinations, it is necessary for clinicians to become familiar with the dosimetric principles and to adopt dose reduction techniques in their clinical practice. This chapter reviewed the basic dosimetry parameters necessary to understand the terms and concepts invariably brought up in any discussion of radiation dose optimization methods. The abstract concept of effective dose should be well understood in order to explain to the patients the relative risks of different medical procedures that involve the use of radiation, e.g., the comparative risks of coronary CT angiography, chest X-rays, and radionuclide perfusion studies.

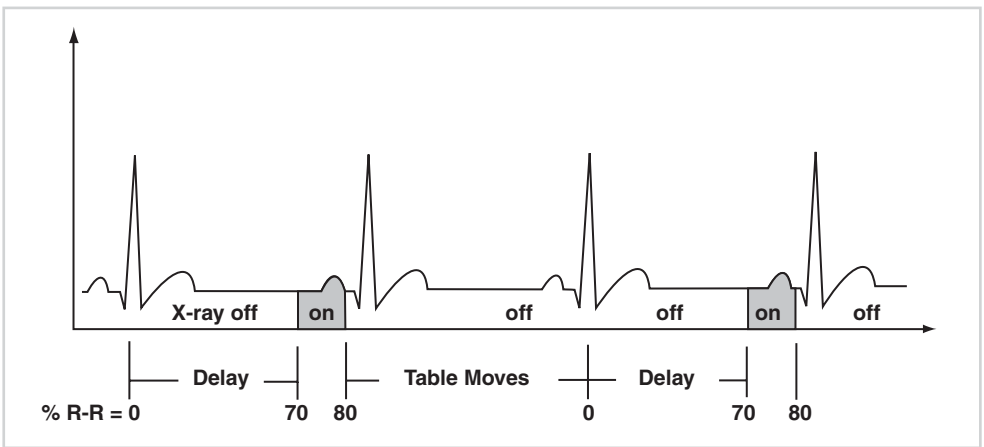

Figure 3.3. Schematic demonstrating prospective gating. When the CT is set to scan, the AEC sets the clock to zero upon receiving an R-wave from the ECG monitor. The tube current is turned off until 70% of the cardiac cycle has elapsed. After this initial delay, the tube current is turned on to the maximum and maintains the maximum output for the next 10% of the cardiac cycle. The AEC subsequently turns off the tube current at 80% of the R–R interval.

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