2 Cineangiographic Imaging, Radiation Safety, and Contrast Agents

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In addition to mastering clinical, pharmacologic, and technical knowledge (see Chapter 1), an interventional cardiologist must also have a good working knowledge of the physical, engineering, and radiobiologic principles underlying fluoroscopic-cineangiographic equipment and radiation safety. This applies to both selection and quality maintenance of complex imaging equipment, as well as to its correct operation to provide optimal images while protecting both patients and staff from unnecessary radiation. Safety is more than a theoretical concern, because severe radiation injuries have occured as a result of prolonged interventional procedures with fluoroscopy times >30 minutes (1–3). The introduction of beta- and gamma-brachytherapy into the catheterization laboratory (see Chapter 22) has further broadened the operator knowledge requirements about radiation biology and radiation safety, and formal documentation of radiation competency is now expected of interventional cardiologists. The qualifying examination for the Certificate of Additional Qualification in Interventional Cardiology thus assigns 15% of the examination to imaging and radiation safety (4). A current intersociety clinical competency statement outlines the necessary body of knowledge (5), with similar guidance available in Europe (6).

This chapter consists of major sections devoted to basic x-ray physics, fluoroscopic imaging technology, radiopathology, patient radiation management, staff radiation safety, and iodinated contrast agents. It is current at the time of its writing, but those seeking more detailed technical information are referred to standard textbooks in the field (7–9).

BASIC X-RAY PHYSICS

X-rays are a form of electromagnetic radiation, like their longer wavelength cousins, radio waves and visible light. Like light, the x-ray beam can also be viewed as a stream of particles (photons, i.e., discrete packets of electromagnetic radiation that each contain a defined amount of energy. By virtue of its very short wavelength and related very high frequency, each x-ray photon contains thousands of times the energy of a photon of visible light. This explains why x-ray photons can pass through solid matter and why different and more potent biologic effects occur when an x-ray photon is absorbed or scattered from living tissue.

X-Ray Dose and Its Measurement

There are many ways to measure radiation units, and a full explanation of all of the current dose definitions, and those of related older units, is available in the literature. This chapter focuses on those dosimetric units that are of importance in the interventional laboratory (Table 2.1).

Exposure is the radiation level at a point in space, commonly measured with an ionization chamber in units of air kerma (*k*inetic *e*nergy *r*eleased in *ma*terial; dose delivered to air). The older unit was the roentgen, R, defined as 2.58×10^{-4} coulombs per kilogram of air. By itself, exposure gives no direct information regarding how much radiation energy is delivered to a person or the biologic effects that irradiation might have.

TABLE 2.1
CLINICALLY IMPORTANT DOSIMETRIC DEFINITIONS

Dose (to a defined substance) Unit is the gray.	The concentration of radiation energy locally absorbed by the defined substance. Under almost all circumstances, the dose delivered by an x-ray beam varies from point to point in the patient. 1 Gy (specific substance) = 1 joule (absorbed)/kg (specific substance)
Air kerma (exposure) Unit is the gray.	The dose delivered to air at a point in space. Fluoroscopic output is usually stated in terms of air dose at a reference point. This value can then be used to calculate patient-related dosimetric quantities. 1 Gy (air) = 1 joule (absorbed)/kg (air)
Effective dose Unit is the sievert.	A calculated quantity based on the physical dose delivered to each of the patient's tissues and modified by the sensitivity of that tissue to cancer induction. It is therefore a measure of the risk of cancer induction caused by that irradiation. Radiation protection guidelines are often expressed in terms of effective dose. ED (Sv) = \ [Dose to a volume of tissue (Gy)] × Radiosensitivity of that tissue
Skin dose Unit is the gray.	The dose delivered to a portion of the patient's skin during a procedure. It is the sum of the dose delivered by the imaging beam and the dose delivered by x-ray photons backscattered from the patient toward the entrance surface. Backscatter adds approximately 30% to the entrance dose delivered by the fluoroscope in typical invasive cardiology settings.
Peak skin dose Unit is the gray.	The maximum dose delivered to any portion of a patient's skin during a fluoroscopic procedure. Deterministic radiation injuries, such as skin burns, are produced if the peak skin dose exceeds a threshold value.
Dose–area product (DAP) Unit is the gray cm^2.	The product of the air dose at a certain distance from the x-ray tube and the cross sectional area of the x-ray beam at the same distance. DAP is actually independent of distance; as distance increases, air dose decreases and beam size increases in an exactly offsetting manner. Because most of the x-ray beam is absorbed by the patient, DAP is a conveniently measurable surrogate for effective dose. Most currently used interventional fluoroscopes include a DAP meter.
Cumulative dose Unit is the gray (air kerma).	The air dose delivered to the interventional reference point during an entire procedure. It includes both fluoroscopic and cinefluorographic contributions to the total. It represents the skin dose from that procedure only if the beam does not move during the procedure.
Dose at the interventional reference point (DIRP) Unit is the gray (air kerma).	The air dose delivered to a defined reference point. This point is selected to be representative of the entrance skin surface of an average sized patient's skin for a fixed x-ray beam. Under these circumstances, DIRP provides a reasonable estimate of peak skin dose.

Dose refers to the local concentration of energy absorbed by tissue from the x-ray beam when the exposure interacts with the individual atoms in the tissue. Specifically, dose is the amount of energy absorbed from the radiation field by a small volume of tissue, divided by the mass of the tissue. This is currently measured in gray (Gy, or 1 joule per kilogram), which corresponds to a very large radiation dose. Accordingly, dose is more often expressed as centigray (cGy, 1/100 Gy), equal to the older unit of dose known as the *rad* or radiation absorbed dose or milligray (mGy, 1/1000 Gy). Because the dose delivered by an x-ray beam is almost always nonuniform (owing to absorption of incoming photons by the more superficial tissues with fewer photons available to deliver dose to deeper tissues), the highest dose is generally delivered to the skin where the beam enters the patient (entry dose). There are also tissue-to-tissue differences in absorbed dose for the same exposure (the principle on which x-ray imaging is based), with water absorbing 0.9 rad (9 mGy) per 1 R of exposure, compared with bone absorbing 4 rad (40 mGy) per 1 R of exposure.

Different types of radiation produce different degrees of damage (i.e., alpha particles versus x-rays), which is taken into account by an absorbed *equivalen*t dose expressed in units of sievert (in older units, the unit was rem [radiation equivalent in man] with 10 mSv equal to 1 rem). Although this distinction is important for certain types of radiation, the distinction between Gy and Sv (or between rad and rem) is largely semantic, because they (Gy and Sv) are essentially equivalent for diagnostic x-rays. In 1987, the National Council on Radiation Protection (NCRP) introduced a new term, the effective dose equivalent (EDE, also measured in Sv or rem), which is a weighted average that takes into account the physical distribution of radiation and the relative radiosensitivity of different organs.

Clinical Measurement of Radiation

Two important locations at which to measure dose are the patient's entrance skin and the entrance to image receptor (Fig. 2.1). Measurements at the patient's entrance surface provide information needed to calculate the radiogenic risk to the patient, whereas measurements at the image receptor entrance define the number of photons available for image formation and determine image noise. Direct

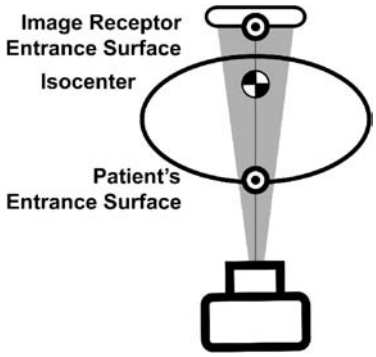

Figure 2.1 Patient and image receptor dose measuring points. Dose measurements referenced to the point at which the x-ray beam enters the patient are useful for estimating patient radiation risks. Measurements referenced to the image receptor entrance can be used to manage image quality.

measurements of patient entrance dose during a procedure are not compatible with clinical routine, so fluoroscopic systems provide a variety of indirect measurements that can be used to estimate actual skin dose. The most basic of these is fluoroscopic time, the time (in minutes) when the x-ray beam is "on" during a procedure. This was a useful tool for the manually controlled fluoroscopes available in the 1950s, but it is of limited value now because it neither keeps track of cine usage nor reflects the effect on patient entrance dose owing to tissue thickness. Most modern cardiovascular fluoroscopes thus incorporate software to estimate the *dose–area product (DAP).* DAP includes both fluoro and cine exposure and reflects the influence of tissue thickness on skin dose. But because the same DAP can be delivered as either a high dose to a small field size or as a low dose to a large field size, it cannot be used directly to predict the possibility of a skin injury (which would be significantly higher in the former case). Rules of thumb, however, do allow DAP meters to be used to manage skin risk with reasonable accuracy, as discussed below.

Interventional systems compliant with the International Electrotechnical Commission's (IEC) standard on Interventional Fluoroscopic Safety (10) also provide a measurement of estimated dose at the entrance surface of a normal-sized patient under typical angiographic conditions. This may overestimate or underestimate skin dose for a real patient and assumes that the beam entry point is constant during a procedure. It thus overestimates actual skin dose when there is considerable beam movement during a procedure, as in performing angiography from different angles.

X-Ray Production

X-rays are produced when high-energy electrons are decelerated by interacting with a metallic target (in our case, tungsten). For that reason, the principal method of x-ray production is usually called *bremsstrahlung* (breaking radiation). The resulting x-ray photons have a spectrum of photon energies, from approximately 20 KeV up to the maximum voltage applied to the x-ray tube (usually 70 to 120 KeV). Some additional x-ray production occurs when the incoming electrons interact with the orbital electrons of the target's atoms. Because the emerging x-ray photons produced by this means carry a particular energy characteristic reflecting the energy levels of the target's atomic orbits, these are called characteristic x-rays.

X-Ray Image Formation

An x-ray beam traveling through a uniform material would carry little information. When the beam travels through tissues with different x-ray absorbance, different fractions of the incident radiation are absorbed (Fig. 2.2); thus the beam leaving the patient is modulated by the pattern of differential absorbance. The modulated beam exits from the patient and is detected by an image receptor. An object can be delineated in the image only if its x-ray absorbance is sufficiently different from that of its surrounding structures to produce sufficiently different exit beam intensity in that location, as a function of the atomic number of the attenuating material, x-ray photon energy, physical density (gm/cc) of the object, and the thickness of the object.

Natural differences in absorbance between tissues can be enhanced by using a contrast agent—a material of markedly different absorbance than the tissue—which can be delivered or concentrated in an anatomic structure of interest. In the catheterization laboratory, the intravascular contrast agents described later in this chapter are based on iodine, whose high atomic number and x-ray spectrum (it absorbs intensely at 40 to 75 KeV) allow visualization of even small (submillimeter) vessels on the x-ray image when the iodine-containing contrast agent displaces lower-density water (blood) during angiography.

Unfortunately, when higher energy x-ray spectra (>100 KeV) are used to image the heart through long tissue paths, the difference between iodine or steel (i.e., a stent or guidewire) and the water density of surrounding tissue is reduced substantially. This is one of the reasons why stents and contrast media vary in visibility from view to view and from patient to patient.

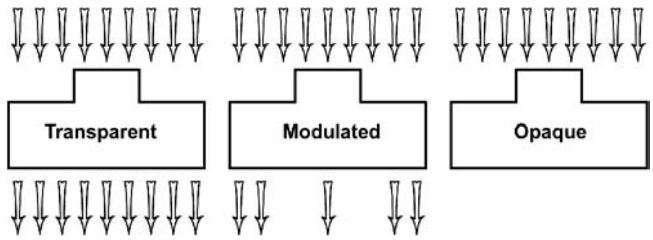

Figure 2.2 Beam modulation. Total transmission of the x-ray beam produces a uniform signal. Total attenuation produces a silhouette. Image formation requires attenuation of a portion of the x-ray beam. Thus patient dose is unavoidable (see reference 5).

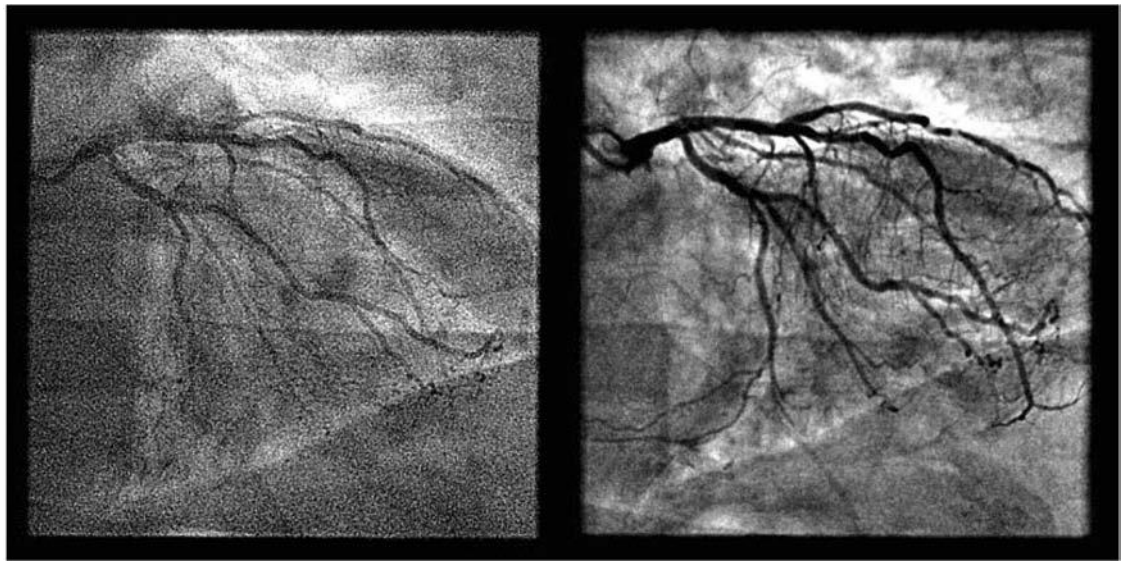

Figure 2.3 Image receptor dose, system settings, and image quality. The image on the left is a fluoroscopic last image hold (LIH). The image on the right is a cine frame. The increased noise in the LIH results from less dose used in its production. The increase contrast in the cine results from system programming to a lower kVp.

Image Noise

A radiographic image of even a uniformly dense object will have random point-to-point variations in brightness over time. These random fluctuations are called *image noise* (commonly referred to as *quantum mottle*). As the x-ray dose striking the image receptor decreases (Fig. 2.3), the amount of image noise increases (fewer imaging photons equals a noisier image). Noise reduces the ability to detect low-con- trast structures, but this can be overcome by increasing the dose, thereby suppressing noise and increasing our ability to resolve such structures. On the other hand, the desire for a low-noise image always must be balanced against the fact that increasing dose also increases patient x-ray exposure.

Scattered Radiation

Scattered radiation is produced when the x-ray beam inter- acts with the patient and is redirected rather than absorbed completely. If scattered radiation reaches the image receptor, it contributes to noise and reduces the image contrast cre- ated as the primary x-ray beam interacts with the anatomic structures. Scattered radiation is also the principal source of exposure for the patient's body parts that lie outside the field of the primary x-ray beam and also for the laboratory staff. The amount of scatter increases with increases in the inten- sity of the x-ray beam and the size of the x-ray field.

Optimizing the Exposure Parameters and Image Quality

From the discussion above, it is clear that the goal of pro- ducing a usable x-ray image requires a number of trade-offs.

Ideal x-ray imaging parameters must appropriately balance the requirements for contrast (needed to detect the object), sharpness (needed to characterize it, including image noise), and patient dose. The dose must be chosen at the minimum level that will generate an image with an accept- able degree of noise, to minimize patient exposure. Increasing *kVp* (using more energetic photons) can pene- trate a large patient more easily and thus reduce patient exposure, but it decreases image contrast significantly. Decreasing the *image receptor input dose* reduces patient exposure, but increases image noise. Thus, for a given patient size, there is an optimal balance that provides acceptable image contrast at an acceptable image noise level while minimizing patient dose. In most modern cine- fluorographic units, programs installed at system setup are designed to give a clinically useful balance between these parameters automatically, although some configurable set- tings can be programmed by the user if flexibility is desired.