Proper Use of Cineangiographic Equipment and Contrast Agents

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Although the training for performing cardiac catheterization procedures includes detailed study of cardiac anatomy, physiology, and pathophysiology, such training usually does not include formal instruction concerning radiologic equipment, radiation safety, or the optimal use of contrast agents. As a result, most operators take a “learn as you go” approach to understanding these increasingly complex areas. Crises may develop when it becomes necessary to select new equipment or to describe malfunctions of existing equipment to service personnel. Even between crises, laboratories run by such operators typically devote little attention either to maintaining optimal image quality or to ensuring the radiation safety of their patients and personnel. Accordingly, this chapter attempts to heighten awareness about the equipment used for cardiac angiography, radiographic principles, programs for radiographic quality assurance, radiation protection, and the characteristics of various intravascular contrast agents. Those seeking more detailed technical information are referred to the 1991 American College of Cardiology/American Heart Association Guidelines for Cardiac Catheterization and Cardiac Catheterization Laboratories and Radiation Safety, as well as the excellent text of Moore.

THE ANGIOGRAPHY ROOM

The modern cardiac catheterization laboratory (Figs. 2.1 and 2.2) consists of a patient support table, equipment for monitoring intracardiac pressures and electrocardiographic activity, and a floor or ceiling-suspended gantry that allows variable angulation of the x-ray beam through the patient. The patient support consists of an adjustable-height, flat-top table whose locks can be released to allow the table top to “pan” freely (i.e., move the patient’s body horizontally toward the head or foot, left or right) within the x-ray beam. The reliance on the gantry to perform all beam angulation represents a distinct improvement (in terms of both patient comfort and maximum achievable angulation) over the cradle systems used in the 1970s. In these earlier systems, the image chain remained stationary while the patient was rolled from side to side to provide views of the heart in different projections. Regardless of its design, the only purpose of the support equipment is to allow precise positioning of the radiographic imaging chain relative to the patient, in terms of both rotation (left or right anterior oblique) and skew (cranial or caudal) angulation (Fig. 2.3). In some laboratories, a second complete imaging chain may be used to provide simultaneous viewing of cardiac structures from a separate angle. Such “biplane” imaging systems are significantly more expensive than “single plane” systems, and are generally preferred only by laboratories that study a high percentage of congenital cases or whose operators believe that biplane imaging is of value in certain procedures (e.g., transseptal puncture, electrophysiology ablations).

FIG. 2.1.

Biplane cineangiography room showing (1) anteroposterior (AP) plane image intensifier, cine camera, television camera, and x-ray tube (partially hidden below table), attached to a floor-mounted parallelogram gantry to allow complex angulation; (2) lateral plane image chain attached to a ceiling-suspended gantry that can be moved into place when biplane imaging is desired; (3) and (4) television monitors for AP and lateral image chains; (5) movable radiation shield to protect operators from scatter dose to eyes and thyroid; (6) physiologic pressure recorder located behind lead-glass window to protect the cardiovascular technician from scatter radiation; (7) remote display of physiologic data; (8) power injector for contrast delivery during ventriculography; (9) emergency cart containing defibrillator, airway management, and drug supplies; and (10) patient support table with (c) control box for table and gantry movement, magnification mode, and beam-restricting “cones” and (j) junction box for connection of pressure transducers to physiologic recorder. The generator and image chain electronics are concealed behind the louvered doors seen along the right wall of the room.
The L/C stand (General Electric, Milwaukee, WI) is a commonly installed alternative to the parallelogram for complex angulation.

Angulations used in cardiac angiography include both rotation and skew. Rotation is shown here for the classic positions in which the image intensifier is at 60° left anterior oblique (LAO) and 30° right anterior oblique (RAO), relative to the vertical, in the transverse plane. Skew is shown here for the caudocranial (now called cranial) angulation, in which the image intensifier is positioned closer to the patient’s head, and the craniocaudal (now called caudal) angulation, in which the image intensifier is positioned closer to the patient’s feet. Complex angulations are specified by giving both the rotation and the skew (e.g., “60° LAO, 35° cranial”).

The classic image chain consists of a generator and cine pulse system, an x-ray tube, an image intensifier, an optical distributor, a 35-mm cine camera, and a television camera and monitor. As such, the image chain can provide both “live” fluoroscopy to facilitate placement of cardiac catheters and cineangiography to permanently capture details about the anatomic and functional state of cardiac chambers, great vessels, and the coronary circulation. Although 35-mm film was the original medium for recording cineangiographic images, since 1998 virtually all newer installations are based on “filmless” technology in which the 35-mm camera is eliminated and the images viewed by the television camera are permanently recorded by digital encoding (see later discussion).

To house this bulky and expensive equipment and the support personnel required to operate it, the room should have a floor area of at least 500 ft² (47 m²) with a ceiling height of at least 10 ft (3 m) (1). The walls should be shielded with 1 mm of lead up to a height of 7 ft to provide radiation protection for personnel in surrounding work areas, and any observation windows into the room should be made of lead-treated glass or plastic that provides a similar level of radiation shielding. Although the control and recording equipment was once situated within the room itself, current room design segregates this equipment and operating personnel within a “control area” that provides a similar degree of lead shielding without sacrificing excellent verbal communication or rapid access to the procedure room itself. The bulky components that constitute the generator and its associated electronics (i.e., the “racks”) may be placed in ventilated closets along the walls of the room but must be positioned so that the high-voltage cable runs are short (less than 40 ft [13 m]) and the racks themselves are easily accessible to service personnel for diagnostic and repair activities.

The Generator

The generator is basically a step-up transformer that converts three-phase 480-V line current into the high voltage (70 to 120 kV) and current (300 to 800 mA) needed to power the x-ray tube for the generation of an x-ray beam (4). The transformer is submerged in a large tank of oil for cooling and insulation. The alternating current (AC) output of this transformer is then converted into direct current (DC) by rectifier circuits.

To be useful in a cardiac study, the DC output of the generator must be combined with a cine pulse system, which chops the generator output into the brief (4- to 6-msec) pulses that are required to “freeze” motion-induced blurring of the rapidly moving coronary arteries. Such pulsing is best performed in the secondary or high-voltage side of the system, using either solid-state or vacuum switching tubes (triodes or tetrodes) or a grid-controlled x-ray tube to transiently interrupt the generator current. The cine pulse system must be capable of handling the 60- to 100-kW power output of the generator and delivering it to the x-ray tube. The generator must also contain automatic brightness control (ABC) circuitry, which allows it to compensate for changes in the transmission of x-rays to the image intensifier that occur as the beam is “panned” through structures of differing attenuation. This is accomplished through the use of a photocell (located at the output of the image intensifier) that detects a drop in light and triggers the ABC to increase the generator output so as to return the number of x-rays striking the photocell (and thus the image intensifier) to the level that is optimal for image production. This increase in generator output may be accomplished by increasing one of three factors: $kV$ or kilovolts, the energy of the x-ray photons (which increases...
their penetrating power); \textit{mA or milliamperes}, the electrical current flowing through the x-ray tube (which increases the number of x-ray photons generated); \textit{msec or milliseconds}, the duration of each x-ray pulse (which increases the amount of time the x-ray beam is on and therefore the total number of photons passing through the patient). How these factors are changed is determined by “regulation curves” programmed into the ABC by the manufacturer.

In a steeply angulated view, the ABC system usually relies on increases in kV, because a comparatively small (6 to 8 kV) rise in kV will augment x-ray penetration, as much as doubling the mA or msec. Although doubling the mA would provide the same increase in film blackening, this might well exceed the power-handling capacity of the cine pulse system or x-ray tube (power = kV \times mA) and would double patient dose. Widening the pulse width would also increase film blackening, but doubling the msec would cause significant motion blurring and would also double patient dose. So alteration in the kV is the primary change made by the automatic exposure system in its effort to maintain optimal film blackening. But increasing kV is not without its price, because imaging of iodine-based contrast agents is best achieved with relatively low tube voltages (70 to 80 kV). The x-rays generated at these low voltages have energies that are only slightly greater than the 33-keV binding energy of K-shell electrons in the iodine atoms, so x-ray absorption is very intense. This maximizes the differential absorption by iodine versus water (and therefore the imaging contrast). An underpowered generator or cine pulse system—one that can achieve adequate penetration only by using higher (100 to 120) kV energies well above the K-shell binding energy—tends to produce grainy, low-contrast angiographic images in which both water and iodine look gray.

The X-Ray Tube

The x-ray tube converts the electrical energy produced by the generator into a stream of x-rays, much as a light bulb converts electrical energy to visible light. The x-ray tube consists of an evacuated glass or metal housing that contains a tungsten filament (housed in a cathodal focusing cup) and an anode disc (tungsten alloy, 100 to 120 mm in diameter) that rotates at more than 10,000 rpm during cineangiography (Fig. 2.4). Electrons boil off the filament by thermionic emission, in proportion to filament temperature. These electrons accelerate toward the anode under the influence of the electric field (approximately 100 kV) supplied by the generator. Their sudden decelerative interaction with the tungsten atoms in the anode results in the emission of x-ray photons by the Bremsstrahlung (braking) reaction. The resulting x-ray photons are emitted at a right angle to the direction of electron travel, and they exit the x-ray tube through a “beam port” in its lead housing. For sharpest imaging, this beam should be as narrow as possible, as though it were coming from a single point source.

FIG. 2.4.

X-ray tube construction. Electrons liberated from the heated filament of the cathode are accelerated toward the slanted surface of the rapidly rotating anode. On impact with the anode, sudden deceleration of the electrons generates x-ray photons, which leave the tube housing through a side-positioned window.

Practically speaking, it is difficult to obtain a true point source of radiation, because concentrating the high-power electron beam on a single point would melt the tungsten anode (3,370°C). To reduce the amount of local heating, the anode is rotated rapidly (10,000 rpm), so that the beam impact point is spread along a track around the anode surface. The impact zone is broadened further by beveling the anode so that it faces the electron beam at a slight angle (Fig. 2.5). The amount of “target angle” is critical: too flat a target angle (less than 8° to 10°) provides little extra area for beam impact and also causes a “heel effect,” whereby the anode itself absorbs part of the generated x-ray beam, so that it fails to illuminate the full surface of a 9-inch field. Too steep a target angle avoids these limitations but causes undesirable broadening of the x-ray beam, with resultant loss of image sharpness. Within the range of acceptable heel angles, the apparent size of the focal spot as seen from the image intensifier is referred to as the “effective focal spot”; it is influenced by both filament geometry and the target angle. Most catheterization laboratory x-ray tubes include two different focal spots. The \textit{small focal spot} (usually 0.6 mm) more closely resembles a point source and thus minimizes geometric unsharpness of the image, but it is quite limited in terms of its power-handling capacity (35 kW). This forces the ABC control to resort to a low mA–high kV technique, which provides adequate film blackening only at the expense of poor image contrast. The small focal spot therefore usually is used only for fluoroscopy or for cineangiography in pediatric patients. In adult patients, routine cineangiography uses the \textit{large focal spot} (usually 1.0 mm). Although there is some loss of image sharpness, the fact that the electron beam is spread...
out over a larger area of the anode gives the large focal spot a higher (100 kW) power-handling capacity, which allows use of a lower-kV technique with better image quality.

FIG. 2.5.

Focal spot geometry. To reduce local heating, the intensity of the electron beam at the anode is decreased by rapid rotation of the anode, which spreads the electron beam into a rectangular shape by beveling the anode surface where the beam hits (target angle). The effective focal spot is defined in the direction perpendicular to the electron beam (out the tube window) and represents the size of the x-ray source as seen from the image intensifier (typically 0.6 or 1 mm). Too much bevel widens one dimension of the effective focal spot excessively, and too little bevel (target angles less than about 10°) causes the anode itself to absorb part of the beam (heel effect), making it impossible to illuminate the full surface of a 9-inch image intensifier.

In addition to the problem of instantaneous power loading of the focal spot (kW), x-ray tubes must be able to absorb a large cumulative heat load. Less than 1% of the electrical energy delivered to the tube is converted to x-rays; the rest is retained as heat. The heat is transmitted to the underlying (usually graphite or molybdenum) anode disk, and from there into the body of the x-ray tube. The heat load can be expressed in terms of heat units (HU = 1.35 × kV × mA × sec), and dissipation of this heat load is one of the biggest challenges of x-ray tube design. A typical single-frame cine exposure delivers roughly 300 HU, and a 10-second run of such exposures at 30 frames per second (fps) delivers 90,000 HU. Most x-ray tubes can absorb only 400,000 HU before their support bearings seize and the tube self-destructs, although some modern tubes that use liquid metal bearings can absorb more than twice that amount before failing. To permit more prolonged studies, the heat input to the anode must be counterbalanced by heat transfer to the oil-filled tube housing, which can absorb roughly 1.5 million HU before reaching a sufficient temperature to rupture its oil seals. This margin can be extended by using more conductive liquid-metal lubrication or by circulating air, water, or oil around the housing to conduct heat away from it more rapidly. Although “waiting for heat units” used to cause delays in laboratories when rapid sequences of cine runs were performed, these improvements in x-ray tube design have largely eliminated this problem.

Although not part of the x-ray tube per se, beam filtration and collimation are important aspects of the radiation beam. The energies of the photons in an x-ray beam are not uniform but are distributed over a range extending up to the generator kV. The lowest energies in this distribution (those less than 20 keV) are too weak to make it through the patient’s body and therefore contribute only to increasing the entry skin dose (and not to image formation). Before it leaves the x-ray tube housing, the beam is passed through a material (usually 2.5- to 3.0-mm thick aluminum) that “hardens” the beam by selectively removing low-energy x-rays. The x-ray beam must also be limited spatially so that only the area seen by the image chain is illuminated. Failure to do so unnecessarily increases both patient x-ray dose and the scatter radiation received by in-room personnel. It also degrades image quality by increasing the number of x-ray photons that strike outside of the image field and are then “scattered” back so that they strike the image intensifier. The beam is therefore “collimated” by positioning thick lead sheets within the tube housing to restrict the beam to the appropriate field size. Further limitation of the beam can be achieved by using moveable lead “cones,” so that only the area captured on the cine frame is illuminated. Finally, many systems include a moveable “gradient” or “wedge” filter that can be positioned over the lung field so as to attenuate the radiation that would otherwise cause excessive brightness there that might interfere with imaging of the adjacent cardiac structures.

The Image Intensifier

Even a precisely generated x-ray beam is useless unless the pattern created as the beam passes through the patient results in a highly detailed “shadow” that the operator can view. In the beginning of the 20th century, this was done by allowing the x-rays to strike a fluorescent screen (zinc cadmium sulfide), where they produced a glow so faint that it could be seen only if the operator’s eyes had been dark-adapted by donning red goggles for 30 minutes before viewing. In the 1950s, this situation improved considerably with the development of image intensifiers (evacuated glass bottles coated internally with a fluorescent phosphor at each end) that increase the brightness of the image more than 1,000-fold (5). Figure 2.6 shows the image intensifier in cross-sectional view. X-rays that have passed through the patient strike the input phosphor (usually cesium iodide), which then emits light. Because the input phosphor is in contact with a photocathode, this light causes the release of low-energy electrons into the interior of the image...
The image intensifier also contains an electrostatic lens that focuses the electrons during their flight. Although single-mode image intensifiers are still available, cardiac catheterization imaging systems can vary the focus potential to change the image magnification (dual- and triple-mode image intensifiers). A 9-inch (23-cm) mode uses virtually the entire input phosphor area and is well suited to studies such as left ventriculography that require imaging of a large area with only a modest degree of spatial resolution. A 6- or 7-inch (15- or 17-cm) mode displays a magnified image of the central portion of the input phosphor and provides optimal coverage area and spatial resolution for coronary angiography. In a triple-mode tube, the 4.5- or 5-inch (11- or 12-cm) mode provides still greater magnification for special procedures such as percutaneous transluminal coronary angioplasty (PTCA). Use of this “mag” mode for routine cineangiography is not advised, because it overtaxes the generator and tube capacity and requires careful panning by the operator if the whole coronary tree is to be imaged during a single contrast injection. The need to maintain light output from a small field of view causes the ABC to call for more radiation, so that although the total patient dose (reflected by the product of dose × area) is not increased in the “mag” mode, the dose to an individual square centimeter of patient skin may approach the threshold for radiation injury during prolonged imaging. Larger field image intensifiers (i.e., 12 or 14 inches) are not needed for cardiac procedures and may limit extreme gantry angulation, but they are used in some laboratories where peripheral angiography is performed so that the vasculature of both legs can be visualized simultaneously.

The qualities of the image intensifier are central to the performance of the image chain. Certain desirable characteristics, such as gain, quantum detection efficiency, spatial resolution, and contrast, tend to be mutually exclusive, and all available image intensifiers involve some performance tradeoffs. Moreover, an image intensifier’s performance degrades rapidly, so that its useful life is only 3 to 5 years in a busy laboratory before loss of gain and contrast necessitate replacement. Despite these limitations, image intensifiers are currently available with on-line resolution in excess of 4 line pairs per millimeter in the 6- to 7-inch mode, and excellent balances of desirable characteristics.

The Television Chain

Real-time viewing of the x-ray image is essential to position catheters, monitor injections, move the patient appropriately during each imaging run, and perform interventional procedures. Since the 1960s, this has been accomplished by placing a television camera so that it can also (along with the 35-mm cine camera) view the output phosphor of the image intensifier (6). When the intent of imaging is only to position a catheter or device or to perform a test injection, the cine camera does not operate, and the generator provides a low dose of radiation that is adequate to create a television image. Although so-called fluoroscopy therefore involves less than 1/100 of the x-ray beam intensity that is used for permanent, high-resolution image recording (cineangiography), prolonged fluoroscopy times (up to 30 minutes in some interventional procedures) can contribute total patient and operator x-ray doses as great as those of shorter but higher-dose cineangiographic runs. When the intent is to record an image of the highest possible resolution on 35-mm film or electronic media, cineangiography is performed, using the higher x-ray doses needed for optimal image definition. To avoid “blinding” the television camera at this significantly higher level of light from the output phosphor of the image intensifier, the output light is split by a partially silvered mirror—typically 80% to 90% to the film camera and 10% to 20% to the television camera. Even during cineangiographic runs, however, it is important to collect television images that can be viewed by the operator in real time, so that the diagnostic and therapeutic decisions required during the course of an interventional procedure can be made even...
before the cine film is developed. In newer filmless systems, these television images are even used for the permanent archive of the procedure. High-quality on-line television images are therefore central in both fluoroscopy and cineangiography. Understandably, the performance of the television chain has become paramount to modern cardiac catheterization.

The type of image tube used varies in different television systems. In contrast to the standard vidicon (which uses an antimony trisulfide target), most catheterization systems use a plumbicon (lead monoxide target), saticon (selenium arsenic tellurium target), or primicon (selenium tellurium arsenic target) to maximize image contrast and minimize image carryover or “lag” so that less than 10% of the image signal carries over into the third video field. Solid-state (charge-coupled device, or CCD) television pickups are now entering service; they have the advantage of one-to-one mapping of each element in the camera to a “pixel” in a digital television image.

Another important variable is the scanning format. Interlaced scan systems (like broadcast television) alternately sweep the even- and odd-numbered lines from the overall raster. This may result in degradation of the television image during 30-fps cineangiography, due to misregistration of anatomic details between the odd and even scan lines, as well as “flicker” caused by collection of one set of lines when the x-ray beam is on and another set when it is off. Use of a progressive scanning television format can overcome these limitations so that all scan lines are acquired in numerical sequence each time the x-ray tube is pulsed. A scan converter then picks off the even- and odd-numbered scan lines for display on a standard interlaced monitor.

The other scan variable is the number of scan lines that make up the image. High-line systems (1,023 or 1,049 horizontal lines, compared with the standard 525-line system) can theoretically improve vertical resolution on the television monitor, but they also increase amplifier bandwidth and therefore can introduce more electronic noise into the displayed image. These limitations of “analog” television were largely overcome by the introduction of “digital” systems in the early 1990s. These systems use high-speed computers to perform on-line processing of the television image (improving image contrast, noise, and edge definition). In addition, digital systems store the images obtained in all cineangiographic runs and allow their playback at variable speed, magnification, and level of contrast enhancement. These resulting images can then be retained as a “road map” to facilitate the positioning of interventional devices, to compare anatomy before and after intervention, and (most recently) to perform on-line measurement of stenosis severity. Because they provide such substantial improvements in the quality and accessibility of television images that can be studied by the operator in the laboratory, these digital systems have replaced the combinations of videotape recorders and “frame grabbers” that were used in the late 1980s for in-lab review of angiographic runs.

Even newer developments in optical electronics may eventually lead to further improvements in system integration and performance. One such example is flat-panel technology (now being explored for radiography and computed tomographic scanning), which has the potential to replace both the image intensifier and the television chain. This technology uses a cesium iodide layer to turn x-ray photons into light, but then passes these photons on directly to a matrix of millions of tiny photodiodes. The analog electrical output of each photodiode is digitized, processed, and permanently recorded in digital form.

**Cine Camera and Associated Optics**

Although in-lab assessment of coronary images has become central to the performance of cardiac catheterization, permanent recording of these images is required for postprocedure review (as by cardiac surgeons), long-term archiving (to compare serial studies in a given patient), and transfer (sending images to another site to guide subsequent care there). Although our laboratory has recently been able to fulfill all of these functions with electronic imaging, the long-term standard for permanent recording has been the recording of angiographic images on 35-mm movie film, at a speed of 30 fps. Faster speeds (e.g., 60 fps) were once used to perform left ventriculography (thus capturing more time points during systolic emptying) but were abandoned because they entailed a 2-fold higher radiation dose. Some laboratories have recorded coronary images at speeds as low as 15 fps to reduce patient and operator radiation dose. Because many catheterization laboratories still use 35-mm film, this section reviews some of the unique considerations associated with this archival medium.

The heart of the 35-mm cine system is the camera. Derived from similar cameras used in the movie industry, the cine
camera must provide smooth, accurate, and reliable film advancement. The advancement of each frame of film actually triggers the generator to produce each cineangiographic x-ray pulse, so that smooth and reliable camera operation is essential. The cine camera views the output phosphor of the image intensifier through an optical system consisting of matched collimator and camera lenses that maximize light transmission and spatial resolution while minimizing image unsharpness caused by veiling glare. The lens is also equipped with an adjustable f-stop and an interchangeable system of different-sized apertures used to restrict light transmission to balance intensifier output, film characteristics, and processing parameters. The focal length of the optical system determines the framing mode-the way in which the round output phosphor is represented on the rectangular cine frame (Fig 2.7). It is most common to employ maximal horizontal overframing, in which the full width of the output phosphor is recorded on the cine frame. The adjustable cones are used to block out any portion of the image intensifier circle that falls above or below the rectangular cine frame. For convenience in positioning the cones, the television monitor is marked with lines that represent the edges of what will be captured on the cine frame. Positioning of the cones is even more important when the optics are set up for total overframing.

**FIG. 2.7.**

Effect of camera optics on framing mode. A low-power lens system on the cine camera allows the film to record the entire surface of the output phosphor but provides a small image that uses only 58% of the film frame. Higher-power lens systems give a larger image that uses more of the film frame but cuts off either the top and bottom or all four edges of the output phosphor (maximal horizontal overframing and total overframing, respectively). (Modified from Friesinger GC, et al. Report of Inter-Society Commission for Heart Disease Resources. *Circulation* 1983;68:893A.)

**Cine Film Selection, Processing, and Viewing**

In film-based systems, the ultimate quality of the recorded image depends almost as much on the selection of the cine film and its processing parameters as on the other elements within the image chain. Cine film emulsions are generally slower than those of standard photographic films, but virtually any cine film can be used in a given system if f-stop, aperture, and processing parameters are selected correctly. The choice of a particular cine film therefore depends more on other important features such as contrast, latitude, and grain size, as well as the price and service support provided by the film distributor.

Film characteristics are best explored by plotting the characteristic curve (called the H and D curve, after Hurter and Driffield), using a sensitometer to perform calibrated test-strip exposures and a densitometer to measure the resulting film optical density (OD) (Fig. 2.8). Relevant parameters include the base-plus-fog (the OD of the film at step 0, before exposure to light) and the relative speed index (the sensitometer step that produces a specified increment in OD above base-plus-fog). Two indices of film contrast are also measured and-the gamma and the average gradient and-which have to do with how steeply the OD rises with increasing exposure. In general, films with lower contrast can recover more diagnostic information because their greater latitude allows distinction of more shades of gray and hence more anatomic detail between the extremes of black and white. Appropriate latitude is obtainable with an average gradient of 1.2 to 1.6, which is preferable to the steep edge gradients produced by some higher-contrast films.

**FIG. 2.8.**

Film characteristic curves (H and D curves), showing the relationship between light exposure (increasing step number on the horizontal axis) and film blackening (increasing optical density on the vertical axis). Measurements are shown for two different films, A and B, including base-plus-fog, relative speed index, and gamma (see text). Film B is less sensitive to light but has a lower contrast and wider latitude.

Note that the characteristics described depend not only on the cine film itself but also on how the film is processed. Important variables include the precise chemistry, temperature, agitation, and immersion time used in the automatic processor. When a new film is brought into the laboratory, all processing and image chain variables (e.g., cine dose, cine camera aperture) must be matched carefully by the technical representatives of the x-ray equipment and film-supplying company. Laboratory personnel must then ensure that these parameters remain stable from day to day, by
performing routine sensitometry and limited densitometry. Measurements should include the base-plus-fog, the OD of a “speed step” (whose OD is known to be near 1.0), and the “contrast index” (the OD difference between the speed step and the next-higher-numbered step). Base-plus-fog should not vary more than 0.02, and speed and contrast index should not vary by more than 0.1 from the initial values. Even slight variations beyond these levels should alert personnel to a potential problem with the processor that must be corrected before clinical films are run, if adequate film quality is to be ensured.

Documented stability of film and processing characteristics allow daily checks on the x-ray equipment itself. If a known attenuator (2.3 mm of copper) is filmed using a consistent geometry and exposure mode (selected magnification, focal spot, pulse width, and camera speed), system stability is confirmed by day-to-day reproducibility of the OD at the center of the frame of the resulting film (which averages 0.90 OD units in most laboratories). By recording, in addition, the kV and mA used by the automatic exposure system during these test exposures, the stability of the generator, x-ray tube, and image intensifier can also be verified. The daily quality control routine may also involve imaging a resolution phantom to monitor image intensifier and camera focus. More detailed assessment of system function should be performed by technical representatives at least twice a year, with correction of adjustments (aperture, processing parameters) and any equipment defects (e.g., dose, image intensifier focus, contrast) by appropriate service personnel. Without these routine surveillance measures, significant deterioration may occur in imaging performance before it becomes apparent to the operators.

Perceived film quality also depends to some extent on the system used to view films. Suitable systems are available for private viewing or projection of an image on a conventional movie screen in a moderately darkened room (i.e., during conferences). The larger format requires use of a high-output illumination system (arc lamp or halogen bulb). Either claw-advance or rotating-prism systems can provide a high-resolution, flicker-free image of film during forward or backward transport at frame rates up to 60 fps. Like all elements of the image chain, film projectors require regular maintenance and cleaning to deliver optimal performance. And of course it is vital to have ready access to both current and previous studies on patients. This means that a cine file room, equipped with a bank of film viewers and housing at least the previous 6 to 8 months of films produced by the institution, must be available close to the cardiac catheterization laboratory. Older films back to 5 years should be housed in either an on-site archive or an off-site storage facility, so that an old film can be located and brought to the in-house file room within 24 to 48 hours.

The “Filmless” System

Given the rapid developments in the television chain, digital image processing to improve noise and contrast, and digital image recording, it was only a matter of time before “filmless” imaging achieved adequate quality to replace cine film as the review, archiving, and transfer medium for cardiac catheterization studies. Even after sufficient image quality was achieved in the mid-1990s, there was a significant “tower of Babel” problem in which each x-ray equipment vendor was theoretically free to design its own format for encoding and recording the resulting images. Although this might satisfy the review function within a single laboratory, it was less satisfactory for archiving (where standards might change over time, and make old studies unreadable) or for interchange (where the originating and receiving laboratory might use different and incompatible recording standards). It was therefore critically important that a single worldwide standard be developed for at least the interchange function.

The Digital Imaging and Communications in Medicine (DICOM) committee was jointly formed by the American College of Cardiology, the American College of Radiology, and the National Electrical Manufacturers Association to set standards for storing and retrieving these digital images as a replacement for 35-mm cineangiographic film. Their 1995 guidelines addressed two issues: (a) the logical format in which digital information should be recorded (now known as DICOM format) and (b) the physical medium on which recording should take place for interchange with other facilities. Although each manufacturer is free to set internal standards relating to storage format and medium, each must be able to produce a copy of the study on a recordable compact disk (CD-R) with 2:1 lossless compression.

Although the DICOM standards made it possible to set up filmless laboratories, the solution of having a file room filled with CD-R boxes instead of cans of cine film offers only partial relief. It is still possible to have a study lost, misfiled, damaged, or simply in use by another physician when it is needed. One key to the long-term success of the
filmless revolution will be the development of networks linking the individual catheterization laboratories and review stations to deep digital archives that can store 5 years' worth of studies and retrieve them quickly and reliably (11). Studies should be available quickly, at any of several viewing stations on a network, and may even be sent over high-bandwidth connections to other institutions. Of course, all such systems currently retain the ability to “burn” a single-patient DICOM-compatible CD-R for physical transfer to an outside viewer who has a personal computer with the needed viewing software.

Even at the lowest acceptable resolution standards of $512 \times 512$ pixels with 8 bits/pixel (256 levels of gray), a single cine frame consists of 256 kilobytes (KB) of data. At 30 fps, a moderately “long” angiographic study of 160 seconds (4,800 frames) would consume 1.2 gigabytes (GB) of uncompressed data. With 2:1 lossless compression, one study would fill a current CD-R disc, and even a small laboratory that performed 1,000 studies per year would require truly huge amounts of digital storage—700 GB, or almost 1 terabyte (TB), to store a year's worth of cases. The problem would be exacerbated further by the use of high-resolution imaging ($1,024 \times 1,024$ pixels at 10 bits/pixel), which increases storage requirements by 16-fold. Although storage jukeboxes of digital tape are being developed that can hold hundreds of terabytes, the other solution is to accept higher levels of data compression. In fact, excellent clinical images can be reconstructed even after “lossy” compression up to 15:1 (12). Although the majority of catheterization laboratories that have been installed since 1998 have been filmless, many of these issues still need to be addressed to solidify the ability of this technology to completely replace all of the functions of cine film.

**RADIATION SAFETY**

Cardiac catheterization delivers one of the highest levels of patient and operator radiation dose of all current diagnostic procedures. Although it remains the gold standard for obtaining diagnostic images of and performing catheter-based interventional procedures on the human heart, the potential hazards of radiation should never be far from the operator's mind (2,13,14). This includes both the risks to the patient and the risks to the operator and support staff. Although the patient receives the bulk of the radiation dose, most patients undergo only a few catheterization studies in their lifetime. The operators, however, have daily exposure with cumulative annual doses that are more than 20 times the normal environmental radiation exposure. Although the magnitude of risk from these levels of exposure is comparatively small, it is fair to say that no level of radiation exposure is completely “safe.” Everyone working in the cardiac catheterization laboratory should therefore become familiar with the units in which radiation exposure are measured, the biologic effects of radiation, and proper use of monitoring and shielding equipment.

**Units of Measurement**

The primary unit of radiation exposure is the roentgen (R), which is defined in terms of the amount of ionization that the beam produces in air. In the newer International System of Units (SI), 1 R is equivalent to $2.58 \times 10^{-4}$ coulombs (C) per kilogram of air. To quantitate tissue effects, the more relevant unit is the absorbed dose (rad, or radiation absorbed dose), which measures how much heating the radiation beam produces in each gram of water (1 rad = 100 ergs/g). In SI units, the rad has been supplanted by the term Gray (Gy): 1 Gy = 1 joule (J) per kilogram = 100 rad. But different tissues absorb different amounts of radiation and the principle behind x-ray imaging! For soft tissues, the absorbed dose generally equals 0.9 rad for every 1 R of exposure; denser tissues such as bone absorb 4 rad for every 1 R. In theory, different types of radiation may produce different degrees of damage (i.e., alpha particles versus x-rays); this is taken into account by an absorbed dose equivalent (rem, or radiation equivalent in man), which is expressed in SI units of Sievert (1 Sv = 100 rem). Although this distinction is important for certain types of radiation, the distinction between rads and rems is largely semantic, because they 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 rem or Sv), which is a weighted average that takes into account the physical distribution of radiation and the relative radiosensitivity of different organs.

**Biologic Effects of Radiation**

The average person receives approximately 200 mrem of radiation per year, about half from natural and half from human-made sources including medical x-rays. In the 1950s, various regulatory bodies (concerned mostly with the safety of workers in the nuclear industry) set the maximal annual occupational dose at 5,000 mrem (5 rem, or 50
mSv). It has been suggested more recently that the maximal annual dose should be reduced to 2,000 mrem and that the cumulative lifetime dose limit should not exceed 1,000 mrem per year of life. Although these figures represent the upper limits of acceptable occupational exposure, the better approach is known as ALARA (As Low As Reasonably Achievable). This is because some radiation risks (e.g., genetic defects, cancer) are stochastic—that is, the severity of the problem stays the same and only the probability of developing the problem increases with increasing radiation dose. These stochastic problems may be caused by even relatively small radiation exposures, although at a very low incidence.

Genetic damage (i.e., mutations that are transmitted to the offspring of an exposed individual) is one of the well-known stochastic risks of radiation. In evaluating the risk, one must bear in mind that such mutations occur spontaneously and cause birth defects in roughly 5% of births. Laboratory studies and analysis of survivors of the Hiroshima and Nagasaki atomic bombings suggest that very large exposures (almost 100 rad, or 100,000 mrem) are required to double the baseline risk of mutation (2),(14). Assuming the routine use of appropriate gonadal shielding, a gonadal dose of 200 mrem (2 mSv) per year under a lead apron would translate to 4 rem over a period of 20 years, which would add only 0.1% (to the 5% baseline) risk of such a birth defect. A similar analysis applies to the risk of occupation-related neoplasm related to radiation exposure. Again, based on data from atomic bomb survivors, the risk of fatal cancer is estimated at 0.04% per rem of exposure. Even without considering the effect of appropriate shielding, the allowable occupational dose of 5 rem per year would add only a 0.2% per year increment (to the background 20% spontaneous incidence) of developing a fatal neoplasm.

Unlike these stochastic risks, the major risks of radiation exposure are deterministic (i.e., their onset requires a certain cumulative absorbed dose, modified to some extent by the rate of dose accumulation). The lens of the eye, like most exposed organs, is fairly resistant to direct injury, but at a threshold dose of 250 to 500 rem it may undergo cataract formation. Even assuming no eye protection, this would amount to more than a 30-year accumulation at the current maximal operator doses of 15 rem/yr to the eye. The skin is even more radioresistant, requiring 200 rem in a single dose to produce redness and 300 rem to produce temporary hair loss. Absent placing one's hands in the primary radiation beam, it is hard to see how even modestly prudent occupational exposure to radiation in the cardiac catheterization laboratory would approach these thresholds for deterministic (nonstochastic) injury.

### Measuring Radiation Exposure

The most important factor controlling radiation exposure is the imaging dose. Cineangiographic systems are usually set to deliver 30 µR per cine frame to the face of the image intensifier in the 6- to 7-inch mode frame. This corresponds to a table-top dose of roughly 40 R/min of cine for the average-sized patient (14). During fluoroscopy, the table-top dose is roughly 10 times lower (2 to 4 R/min). The patient's dose comes from his or her position within the direct radiation beam, and the patient's body absorbs most of the incident radiation beam. Less than 1% of the beam actually passes through the patient to strike the image intensifier. Total skin dose to the patient's back can approach 50 R (roughly 50,000 mrem) during the typical examination and radiation-the equivalent of 100 to 250 chest x-ray exposures. Measurements suggest that patient exposures can be substantially higher (approximately 300 R) during complex interventional procedures (15). However, because this dose is distributed among multiple beam entry points as angulation is changed during the procedure, it is tolerable for maximal local skin entry doses not to be exceeded. Prolonged fluoroscopy (i.e., 1.7 hours of continuous fluoroscopy) in a single-beam position can approach the threshold for skin erythema (2 Gy, or 200 R) (16),(17). Episodes of erythema or delayed skin necrosis have been reported in rare instances when prolonged fluoroscopic examinations were performed using a single-beam entry angle and position. In younger patients, there may also be concern about neoplasm or gonadal injury. Dose measurements made during prolonged electrophysiologic ablations suggest a roughly 0.03% increase of lifetime risk of a fatal malignancy associated with undergoing such a procedure. Gonadal dose from secondary scatter is minimal but should be decreased further by avoidance of imaging over the pelvis and use of a pelvic or gonadal shield in younger patients.

Although patient dose is clearly an issue, patients rarely undergo more than a few invasive procedures per year. On the other hand, most operators and support staff perform hundreds of procedures per year. Their main hazard derives not from the portion of the beam that passes to the image intensifier, nor the portion absorbed by the patient, but the remaining portion of the incident beam that is scattered back into the procedure room. At 1 m from the point at which the beam strikes the patient, the scatter exposure is roughly 1/1000 of the patient's skin exposure (i.e., 2 to 4
Because scatter is highly asymmetric (with greatest scatter back toward the x-ray tube from the point at which the beam enters the patient), this is at best a rough estimate. In fact, operator exposure is several times higher for imaging performed in the left anterior oblique or left lateral view, compared with the straight anteroposterior (AP) projection (18). During a standard catheterization, the head of the primary operator would (without other shielding) receive a dose of approximately 20 mrem, about half from fluoroscopy and half from cineangiography (2),(14). At that rate, an operator could perform no more than 250 procedures per year without exceeding the federal guideline of 5,000 mrem/yr. Performance of more complex procedures (e.g., PTCA) or newer device interventions involve greater fluoroscopy time (roughly 25 minutes) (15),(19) and greater operator radiation dose (50 to 100 mrem), which would further reduce the number of procedures that a single operator could perform each year. It should be noted that these doses apply to the first operator, who is standing nearest to the x-ray beam. For assistants who stand further from the beam entry point than the primary operator, radiation exposures are only 0.5 mrem per procedure, or 10% to 30% of the primary operator exposure. For nursing or technical personnel who stand only a few feet further from the patient, doses are even lower, at 2% to 11% than received by the primary operator (2). These doses can be reduced even further by maintaining greater distance when not performing clinically necessary activities that require closer approach or by using portable shielding.

Because the operator dose may be close to the maximal allowable limit and because the dose varies so much depending on geometry, procedure time, and other variables, there is no substitute for having each operator measure his or her own exposure (20). The most common way to do this is by means of a film badge (a small piece of unexposed film in a plastic holder), which must be worn at all times when working in the cardiac catheterization laboratory. The amount of film darkening revealed when the badge is developed reflects the cumulative radiation dose received, with a sensitivity of 10 mrem. If a single “collar” badge is worn, it should be worn on the left shirt collar outside the lead apron, to give an estimate of maximal head and neck exposure. Current recommendations also call for a second “waist” badge, which is worn on the operator's belt just beneath the lead apron. This typically records a dose of only 1 to 2 mrem per procedure, reflecting the lower level of whole-body and gonadal radiation exposure. These two badges should be of different colors (e.g., red for the collar and green for the waist) so that they can be used together to calculate the EDE received, which is equal to 0.3 times the collar badge dose (when worn alone) or to 1.5 times the waist badge dose plus 0.04 times the collar badge dose (when both badges are worn). Each month, these badges should be turned in for processing and replaced with a fresh film packet. Each individual (and the laboratory director) should then confirm that no worker's collar badge dose exceeded 100 mrem/mo without further investigation. The ease of use of film badges has made them the preferred method of monitoring dose, although other techniques (e.g., electronic dosimeters, thermoluminescent salt dosimeters) can be used. The most important issue is that all laboratory personnel must wear their monitoring devices consistently and that their recorded doses must be studied regularly to ensure that occupational exposures are remaining within the prescribed limits.

**Reducing Exposure Dose**

The operator can use several methods to reduce scatter dose (2),(14). The most important of these is minimizing the patient dose, which is the ultimate source of scatter to the operator. This means having the x-ray equipment doses calibrated within national standards and as low as possible without degrading the quality of the fluoroscopic and film images. The temptation to use “high-dose” fluoroscopy (more than 10 R/min table-top dose) should be avoided in all but certain angioplasty settings, when it takes the place of even higher-dose cineangiographic runs. If “pulsed fluoroscopy” is available (in which the x-ray tube current is pulsed during fluoroscopy, as is done routinely during cineangiography), it may improve image quality while reducing patient and operator dose by up to 60%. The operator must also remember to keep the face of the image intensifier in near contact with the patient's chest. Raising the image intensifier off the chest is the equivalent of switching to a higher magnification mode; it increases the radiation output required to provide adequate light output from the image intensifier. It also decreases the ability of the image intensifier housing to screen radiation scattered by the patient before it strikes the operator.

One of the most important means of reducing radiation exposure is reducing the amount of fluoroscopy time to the minimum required to position catheters. The total fluoroscopy time should be recorded in each instance and should be well under 10 minutes for a diagnostic procedure. If a trainee experiences difficulty in accomplishing some task (e.g., advancing the right heart catheter), a more experienced operator should take over rather than allowing...
fluoroscopic time to escalate. It is also important to avoid the “lead foot” syndrome: the operator must learn to depress the fluoroscopy pedal briefly when it is necessary to confirm a catheter position, and to reflexively release the pedal when looking away from the television monitor. Similarly, cineangiographic runs should be selected carefully to show important findings, and each run should be terminated as soon as the necessary information is recorded. Every effort should be made to confine the area irradiated to that imaged, by closure of the adjustable cones to the edges of the cine frame.

The other cardinal measures used to reduce operator x-ray dose are increasing distance and using shielding. The operator should stand as far from the beam as possible to take advantage of the inverse square law—one or two steps further away from the x-ray tube may cut the dose in half. This is particularly important during angulated shots such as the left lateral or LAO cranial projections, which place the operator in close proximity to the beam entry point (Fig. 2.9). Keeping one's distance is easier with the femoral than the brachial approach; the latter carries about twice the exposure to the operator's head.

FIG. 2.9.

Isoexposure curves of operator radiation exposure during cardiac catheterization for U-arm and C-arm systems. The upper panel represents a 30° LAO view; the lower panel shows a 30° RAO view. The operator is standing to the patient's right, and the patient's feet are toward the reader. Radiation is highest at the level of the table and patient, owing to radiation scatter. (Balter S, Sones FM Jr, Brancato R. Radiation exposure to the operator performing cardiac angiography with U-arm systems. Circulation 1978; 58:925, with permission.)

It is also advisable to use additional shielding to reduce exposure of radiation-sensitive areas, analogous to the use of a lead apron to shield the torso. Shielding equivalent to 0.5 mm of lead attenuates exposure by a factor of almost 30 (i.e., to less than 1 mrem per procedure). If the operator has occasion to turn away from the patient, a wrap-around apron should be used to provide protection from that direction. Improvements in two-piece (skirt and vest) and one-piece aprons (with elastic belts to distribute much of the weight to the operator's waist) have made wearing adequate shielding more tolerable. Additional radiation protection can be gained from wearing separate thyroid collars and wrap-around leded eyeglasses. Our preference, however, has been for a movable rectangular lead-acrylic shield (0.5 mm lead-equivalent, available from a variety of manufacturers) that can be positioned between the point at which the x-ray beam enters the patient's body and the operator's head (Fig. 2.1). This shield protects both thyroid and eye as effectively as separate shields, is less cumbersome, and can be covered with a sterile plastic bag to allow placement within the sterile field. Since we began to use such a shield routinely in 1987, not even our busiest operators have exceeded a collar badge reading of 100 mrem/mo.

INTRAVASCULAR CONTRAST AGENTS

Shortly after the classic papers by Roentgen in the 1890s, the search began for effective and nontoxic contrast agents to define vascular anatomy. Although early experimentation involved a number of heavy metals (bismuth, barium, thorium), all modern contrast agents are based exclusively on iodine, which by virtue of its high atomic number and chemical versatility has proved to be an excellent agent for intravascular opacification. Because inorganic iodine (sodium iodide) causes marked toxic reactions, experiments were performed in 1929 using an organic iodide preparation (Selectan) that contained one iodine atom per benzoic acid ring. In the 1950s, a series of substituted triiodobenzoic acid derivatives were developed, which contain three iodine atoms per ring. These agents differ from each other in terms of the specific side chains used in positions 1, 3, and 5 (Fig. 2.10), which influence both solubility and toxicity.

FIG. 2.10.

Sample structures and properties of currently available contrast agents. The traditional high-osmolar ionic (HOCM or ratio-1.5) contrast media are sodium (Na+)/meglumine salts of substituted triiodobenzoic acid that have three iodine atoms per anion-cation pair, with six times the osmolality of blood. Two types of low-osmolality (LOCM or ratio-3)
contrast media are also shown: the true nonionic agents and the Na/meglumine salt of an ionic dimer, which have three iodine atoms per nonionic molecule or six iodine atoms per anion-cation pair, with an osmolality two to three times that of blood. The newest class of isoosmolar (IOCM or ratio-6) contrast medium is a nonionic dimer with six iodine atoms per molecule and an osmolality equal to that of blood. Also included are the iodine contents (in mgI/mL), the osmolality (in mOsm/kg-H₂O), and the viscosity at 37°C. The asterisk (*) indicates a mixed sodium and meglumine salt. See text for details.

Ratio-1.5 ionic compounds are substituted ionic triiodobenzoic acid derivatives that contain three atoms of iodine for every two ions (i.e., the substituted benzoic acid ring and accompanying cation). Included in this family are widely used contrast agents such as Renografin (Bracco), Hypaque (Nycomed), and Angiovis (Berlex), which are mixtures of the meglumine and sodium salts of diatrizoic acid. Functionally similar agents are based on iothalmic acid (Conray, Mallinckrodt) or metrizoic acid (Isoopaque). These agents have a sodium concentration roughly equal to that of blood, pH titrated between 6.0 and 7.0, and a low concentration (0.1 to 0.2 mg/mL) of calcium disodium ethylenediamine tetraacetic acid (EDTA). Higher or lower sodium concentrations may contribute to ventricular arrhythmias during coronary injection. Some suggest that calcium binding by sodium citrate can cause greater myocardial depression (21). To have an iodine concentration of 320 to 370 mgI/mL, as is required for left ventricular and coronary contrast injections, solutions of these agents are markedly hypertonic (with an osmolality exceeding 1,500 mOsm/kg, roughly six times that of blood).

One modification of ionic contrast came in the mid-1980s with the introduction of ratio-3 low-osmolar contrast materials (LOCM). Although it is still ionic (as a mixture of meglumine and sodium salts), ioxglate (Hexabrix, Mallinckrodt) is a ratio-3 agent by virtue of its unique dimeric structure, which includes six molecules of iodine (three atoms of iodine for every one ion). To achieve an iodine concentration of 320 mgI/mL, Hexabrix has an osmolality roughly twice that of blood. This significantly reduces the undesirable side effects related to hypertonicity (22).

A more significant modification in the late 1980s was the introduction of true nonionic, ratio-3 contrast agents. These LOCMs are water soluble in a noncharged form, without an associated cation. Examples include iopamidol (Isovue, Bracco), iohexol (Omnipaque, Nycomed), metrizamide (Amipaque, Winthrop), ioversol (Optiray, Mallinckrodt), and ioxilan (Oxilan, Cook), each of which contain three atoms of iodine for every molecule (23). With calcium disodium EDTA as a stabilizer and tromethamine (1.2 to 3.6 mg/mL) as a buffer, an iodine content of 320 to 370 mgI/mL can be achieved with an osmolality of 600 to 700 mOsm/kg, between twice and three times that of blood. The viscosity of these agents (which influences ease of injection through small-lumen catheters) is roughly six to ten times that of water. More recently, a nonionic dimeric compound (iodixanol [Visipaque, Nycomed]) has been released. This ratio-6 agent actually requires the addition of sodium and calcium chloride to bring its osmolality up to that of blood (290 mOsm/kg) (24).

The ratio-3 LOCMs definitely are better tolerated by patients undergoing coronary angiography (Table 2.1). They produce fewer episodes of bradycardia and hypotension, precipitate less angina, and cause less nausea and sensation of heat than traditional high-osmolar contrast agents (25-27). There is also evidence that the nonionic ratio-3 agents produce fewer allergic side effects and may be less nephrotoxic in animal studies (23), although this has been difficult to confirm in human studies (28). For all of these reasons, more than 80% of coronary angiography studies during the 1990s were performed with an LOCM. Some early studies, however, suggested that the true nonionic agents might predispose patients to thrombotic events (29). This has been one of the strongest marketing claims for Hexabrix, the only ionic LOCM. It is likely that any prothrombotic effects of the nonionic contrast agents reflect more the absence of the antithrombotic properties seen with the ionic agents. In our laboratory, we have not seen practical clinical problems that related to use of nonionic contrast agents. This is consistent with more recent studies (30), which have failed to confirm any increase in deleterious thrombotic complications.

This leaves expense as the major factor that limits the universal use of these agents. Until recently, the 200 mL of contrast material consumed in the average diagnostic catheterization would have cost $150 to $200 for a nonionic agent, compared with roughly $20 for the same amount of an ionic contrast agent. Total conversion to LOCMs therefore would have added $100 to 200 million to the U.S. health care budget. Although randomized trials comparing high- and low-osmolar agents in routine angiography have shown a clear reduction in minor side effects, they have failed to show any significant net clinical benefit in terms of serious side effects that would justify across-
the-board use of a more expensive agent. It therefore has been suggested that most of the benefit of these agents could be secured if their use were confined to those patients who most need their benefits—the roughly 25% of patients who have two or more of the following characteristics: age greater than 65 years, left ventricular end-diastolic pressure greater than 15 mm Hg, New York Heart Association class IV symptoms, or a history of previous reaction to contrast material (25). This is consistent with the practice in our laboratory, where we have used Renografin as our routine contrast agent but have resorted to a nonionic LOCM in roughly 35% of cases based on specific indications: severe hemodynamic dysfunction not responding to pharmacotherapy, history of prior allergic reaction to ionic contrast, internal mammary injection, or baseline renal insufficiency with a creatinine concentration higher than 2.5 mg/dL. With this strategy, the incidence of severe contrast reaction (wheezing, prolonged hypotension, or frank anaphylactoid reaction) remains well below 0.05%.

With the increasing competition among nonionic contrast agents, there has been a marked reduction in price, so that 200 mL of some nonionic LOCMs now costs as little as $36 (two to three times more than a high-osmolar ionic agent). At such a low incremental cost, the clear reduction in minor side effects may be sufficient to justify more liberal use of nonionic LOCMs.