THE OPERATION OF A CINEFLUOROGRAPHIC SYSTEM

The main functions of an x-ray cinefluorographic system are to produce a collimated x-ray beam of appropriate intensity and quality, to project that beam through the patient at a desired angle, to detect the modulated x-ray beam after it passes through the patient, and to transduce the modulated x-ray beam into a usable visible light image. These components are schematically illustrated in Fig. 2.4.

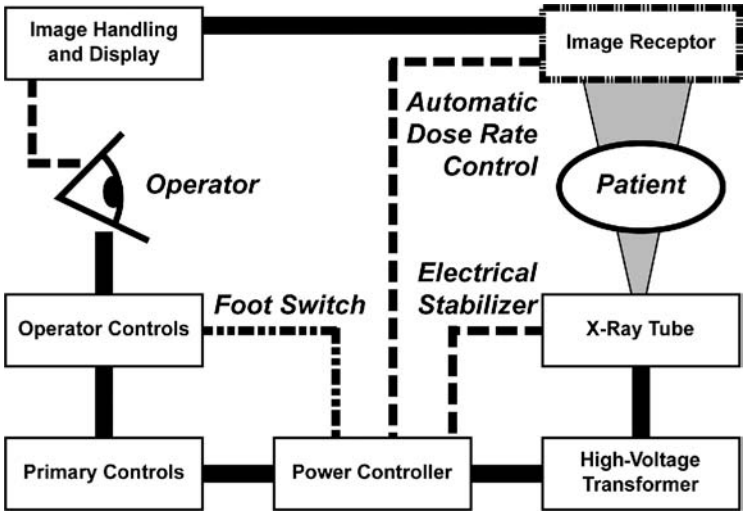

Figure 2.4 Generator and feedback schematic. Modern generators rely on a large number of feedback loops to manage radiation dose and stabilize image quality. The operator provides the final control loop by managing system resources.

Radiation Production and Control

Generators

The cinefluorographic x-ray generator controls and delivers electrical power to the x-ray tube. It heats the x-ray tube's filament to produce a beam of electrons at a current between 1 and 1,000 milliamperes (mA). These electrons are accelerated toward the target anode by high voltage between 40 and 125 DC kilovolts (kVp), applied by the generator. The flow of high-energy electrons toward the anode is not continuous, but is separated into pulses whose duration ranges from 1 to 10 milliseconds (mS) and whose repetition rate ranges from 15 to 60 pulses per second.

Most new x-ray generators use the incoming electrical power to drive a power oscillator operating in the audio frequency range. Voltage is increased by means of a step-up transformer, and the output is rectified (converted to DC) and smoothed before being applied to the x-ray tube. A second circuit supplies a nominal 10 volts to heat the filament of the x-ray tube. Switching circuits effectively turn the electron flow through the x-ray tube on or off to provide beam pulsing. The operator instructs the generator to initiate beam generation at either fluoroscopic or cinefluorographic levels through a pair of foot pedals.

Although beam pulsing was always part of cineangiography to minimize motion and lack of sharpness, earlier systems used continuous x-ray generation during fluoroscopy. In contrast, current digital systems use *pulsed fluoroscopy*, in which the x-ray beam is briefly turned on (pulsed for 1 to 10 mS) once during the recording of each video frame. The fluoroscope must deliver enough radiation dose during each pulse to ensure appropriate image quality, with the video detector storing that image until it is read out and displayed. Digital systems provide gap-filling images to eliminate visual flicker that would otherwise result from frame rates of 15 images per second. Although higher rates may be needed when tachycardia is present, they clearly increase patient dose.

Juggling and optimizing these various x-ray parameters is necessary as angles change and as the beam is moved (panned) across the heart. This is well beyond the manual capabilities of a technologist and therefore requires circuitry in the generator that continuously measures the actual voltage across the x-ray tube, the current flowing through it, the pulse width, and the amount of light generated at the image receptor. These data are used to adjust the input parameters—voltage (kVp), current (mA), and pulse width (mS)—for proper operation.

X-Ray Tubes

The x-ray tube is a device that converts a portion of the electrical energy delivered by the generator into x-rays. The x-ray tube consists of an evacuated glass or metal housing that contains a tungsten filament (housed in a focusing cup), and an anode disc (tungsten alloy, 100 to 200 mm in diameter), which rotates at more than 10,000 rpm (Fig. 2.5). Electrons are emitted from the filament by thermionic emission. The number of emitted electrons, and thereby the tube current (mA), is controlled by adjusting the filament temperature. These electrons accelerate toward the anode under the influence of the electric field (~100 kV) supplied by the generator. The sudden deceleration caused by interactions with the tungsten atoms in the anode produces x-ray photons by the bremsstrahlung (braking) process, as described above. For sharpest imaging, the point of impact of the electron beam on the target should be as small as possible, so that x-ray emission appears to come from a single "point" focal spot.

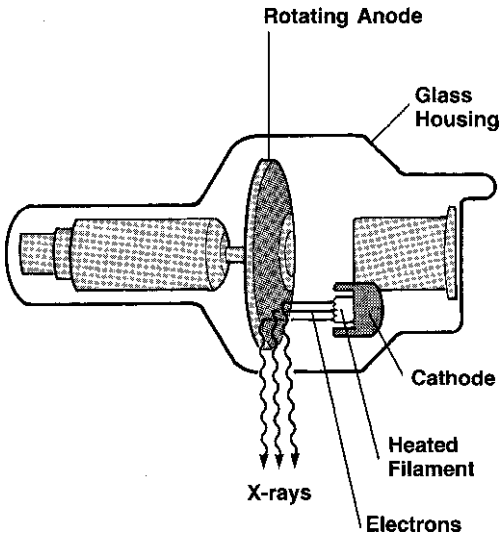

Figure 2.5 Rotating anode x-ray tube. See text for discussion.

The actual size of the focal spot represents a balance between the requirements for sharp and fast imaging and the need to avoid melting the target. X-ray tubes have two filaments and hence two focal spots. The smaller (typically 0.3 to 0.5 mm) is used for fluoroscopy. A larger focal spot (typically 0.8 to 1.0 mm) is used to accommodate the higher power requirements of adult cine. In addition, the anodes rotate at high speed to spread the heating over a long focal track instead of concentrating it on a small point.

X-ray generation is extremely inefficient from the standpoint of energy transformation. Less than 1% of the electrical energy applied to the tube is converted to x-rays. The remainder is deposited in the tube as heat. This creates an important heat dissipation challenge for X-ray tube design. Too much heat delivered in too short a time will melt the anode. Present tube designs are capable of dissipating several times as much heat as those of the early 1990s. Thus, these tubes can deliver significantly more radiation to patients without overload than was possible a decade or so ago. A tube heat warning occurring during a procedure is a clinical indicator that substantial amounts of radiation may have been delivered to the patient.

X-Ray Beam Filtration and Shaping

Although the maximum x-ray photon energy is set by the acceleration voltage supplied by the generator (kVp), the beam contains a spectrum of lower energies as well. These low-energy x-ray photons are easily absorbed by the patient's superficial tissues and thus do not contribute to image formation. To avoid nonproductive entry site exposure, it is good practice to remove (filter) these low-energy photons from the beam before they enter the patient. An aluminum plate placed in the x-ray tube's beam port preferentially absorbs the low-energy photons and increases the effective penetrating power of the resulting beam (beam hardening). Many modern interventional fluoroscopes offer a copper filter as well as the required aluminum filter, because the higher atomic number of copper relative to aluminum produces even more beam hardening than does aluminum. High-power x-ray tubes, copper filters, and appropriate fluoroscopic system programming can be combined to produce an x-ray spectrum that has a large fraction of its photons just beyond the K-absorption edge of iodine (~40 KeV), to provide significant skin dose reduction without adversely affecting iodine visibility (11).

But too much beam filtration can be a liability by discarding too much of the generated x-ray beam. The x-ray tube and generator have clear limits on input power delivery that make it impossible to overcome excessive filtering by increased beam generation. Copper filters of any appreciable thickness are thus too attenuating for adult cine and are automatically removed when the cine pedal is selected. In some systems, the automatic dose rate control system may automatically add and remove filters during fluoroscopy, depending on beam angulation and the patient size. Small changes in path length can thus result in large changes in skin dose rate if one is working near such a transition point. Operators should be aware of the filter operating strategy for the systems and clinical modes in which they work.

The x-ray beam is spatially limited so that only the field of view (FOV) seen by the operator on the monitor is irradiated. Absolute beam limitation requirements are specified in national regulations and international standards. The primary beam port is equipped with lead shutters, which are adjusted automatically as the system tracks the active FOV and distance between the focal spot and the image receptor, to adjust these shutters to appropriately limit the irradiated area. The functionality of this device needs to be checked periodically. The lead shutters can also be manually closed to less than the full FOV. Such collimating of the beam has a beneficial effect on image quality (reducing scatter) while simultaneously reducing both patient and staff irradiation.

Many systems also have movable semitransparent copper shutters (also called *wedges*) that can be positioned over the lung field up to the heart border in each projection to improve overall image quality by reducing excessive image brightness over the lungs. These shutters also help minimize unnecessary patient and staff irradiation.

Imaging Modes

X-ray cinefluorographic units operate in two modes: fluoroscopy and acquisition (cine or image recording). The purposes and x-ray generator operating parameters of the two modes are different, particularly in terms of the input x-ray dose delivered and in image quality. Figure 2.3 shows single-frame images acquired at fluoroscopic and acquisition doses.

Fluoroscopy

Fluoroscopy provides a real-time x-ray image with adequate quality for guiding manipulations. The physiology of vision effectively integrates several frames; this reduces perceived image noise, so greater image noise can be tolerated allowing fluoroscopic x-ray input dose rates that are significantly lower than those used for acquisition.

Current fluoroscopic systems have two or more operator selectable fluoroscopic dose rates. The higher dose rates provide less image noise at the cost of greater patient and operator exposure. Many systems offer variable fluoroscopic frame rates. Decreasing the frame rate saves dose at the expense of visual smoothness of the transition between frames. Because of persistence of vision effects, lowering frame rates does not linearly lower required dose rates (12) Many operators find 15 frames per second (fps) to be satisfactory for digital cardiac fluoroscopy.

Acquisition (Cine)

The acquisition mode generates images of sufficient quality for single-frame viewing. Higher x-ray input dose rates are needed to reduce image noise and optimize clinical visualization, and most x-ray cinefluorographic units are calibrated such that the per-frame dose for acquisition is approximately 15 times greater than for fluoroscopy. A single frame acquired in acquisition mode thus delivers about the same patient dose as one second of pulsed fluoroscopy at 15 fps.

The optimal acquisition mode input dose per frame is that which achieves the best balance between image noise and image quality. The cine dose rate is also directly proportional to the acquisition frame rate. As with fluoroscopy, digital gap-fill can achieve flicker-free image displays at any frame rate, but the image presentation may become increasingly jerky at frame rates below 15 fps despite such gap-fill. The typical acquisition frame rate for adult studies is 15 fps.

Feedback

The x-ray beam is attenuated as it passes through tissue. The degree of attenuation varies with tissue density and other factors such as the projection angle and the distance between the x-ray tube and the image receptor. Feedback circuits measure the brightness of the image generated by the image receptor. This feedback signal is used to modulate the output of the generator in response to changes in patient density and position. This is accomplished by an automatic dose rate control (ADRC) circuit that is designed to maintain a constant brightness level of the image-intensifier output signal. The normal function of this circuit has a profound influence on patient skin dose. X-ray intensity is increased if the detector measures too dim a signal and decreased if the signal is too bright. This means that the patient entrance port skin dose increases substantially when compound projection angles with cranial or caudal skewing are used (Fig. 2.6).

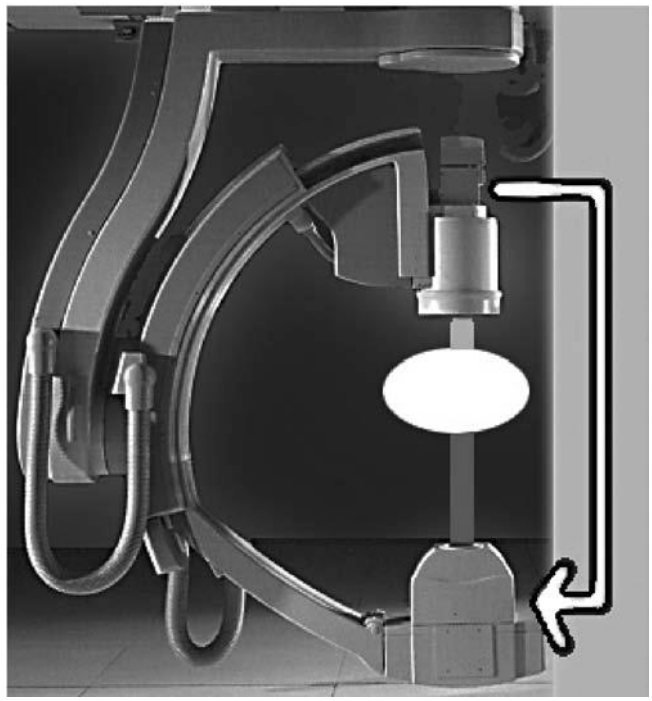

Figure 2.6 Patient size profoundly affects patient dose. Fluoroscopic systems continuously adjust x-ray output to account for differences in patient size and beam angulation. This is accomplished by using feedback circuits that maintain constant image receptor dose by adjusting the x-ray tube's electrical inputs. Skin dose will double for every 4- to 6-cm increase in path length through the patient. (Photo of gantry courtesy of Philips.)

The ADRC can control the tube voltage (kV), tube current (mA), pulse width (expressed in mS), and beam filtration. Different makes and models of fluoroscopes are likely to have different ADRC strategies. Most machines offer different ADRC modes of operation. For example: When the system is set to cine coronary arteries, the ADRC remains functional throughout the entire cine run; when the same system is set to LV lock, however, the ADRC establishes a level early in the run and then maintains that level during the contrast injection phase of the ventriculogram.

X-Ray Detection and Recording

Image Detection

The x-ray image formed by the interaction of the x-ray beam and the patient must be detected and transformed into a visible format. The fluorescent screen was the original x-ray detector used by Roentgen. It was the only fluoroscopic detector available from the discovery of x-rays in 1895 until the development of the x-ray image intensifier in the 1950s. The image intensifier was the enabling technology for coronary angiography (13). At the start of the new millennium, solid-state detectors are now beginning to replace the conventional image intensifier.

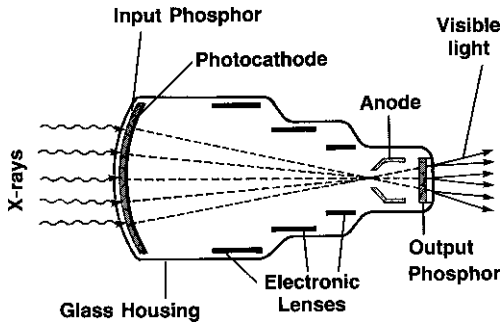

Figure 2.7 X-ray image intensifier. See text for discussion.

Image Intensifier

The structure of a single-mode image intensifier is shown in Fig. 2.7. The modulated x-ray beam emerging from the patient enters the image intensifier and is detected by a cesium iodide (CsI) fluorescent layer. The visible light image emerging from the CsI is converted into an electron image by a photocathode. Focusing electrodes in the tube accelerate and converge the electrons onto a small output window. The electron image is then converted back into a visible light image when these electrons interact with an output screen. The combination of acceleration of the electrons and minifying of the output image relative to the input produces the 100,000-fold brightness gain provided by the image intensifier.

Cardiac image intensifiers offer several magnification modes. These tubes contain a separate set of focusing electrodes for each magnification mode. When a specific magnification is selected, the corresponding set of electrodes is energized. This focuses a larger or smaller portion of the input screen onto the fixed size output screen. The minifying gain of the tube (ratio of active input screen area to the fixed output screen area) decreases as the tube is zoomed; therefore smaller fields of view require higher input dose rates than do larger fields of view.

Cardiac image intensifiers have a typical maximum physical FOV of 23 to 25 cm. For any magnification, the actual FOV can be somewhat smaller if the optics are set up to overframe the image. Smaller FOVs (typically around 17 and 12 cm) provide better spatial resolution at the expense of requiring an increased dose rate. Patient dose can be minimized by working at the largest FOV consistent with appropriately seeing the structures of image. Visibility can sometimes be improved in heavy patients by increasing the FOV and collimating the beam because of the increased minifying gain associated with large FOVs.

Vascular image intensifiers are available with FOVs exceeding 40 cm. When these tubes are operated using a typical 17-cm cardiac FOV, they require a significantly higher dose rate than smaller cardiac image intensifiers. In addition, the larger size of vascular tubes limits beam angulation. Moving the image intensifier farther from the patient to obtain the necessary angles further increases patient dose.

Image intensifiers degrade over time. Service adjustments can compensate for these losses. Eventually the brightness gain deteriorates such that x-ray dose rates must be increased simply to obtain adequate brightness. This will often happen between 3 and 10 years after installation. When the service engineer informs you that the system is at the limits of its adjustment range, it is time to replace the image intensifier.

Cine Camera and Associated Optics

The technology for recording coronary images has almost totally migrated from the cine-film camera to the digital domain. However, film-based technology still merits a brief review both to provide an understanding of older systems and to demonstrate the migration of imaging requirements from film to digital.

Cine cameras are electronically synchronized with the hospital's AC power supply. The usual adult filming speed was 30 fps in the United States and 25 fps in Europe. Most of the cine-film systems built in the late 20th century incorporated pulsed x-ray beams. These systems were programmed to produce x-ray pulse widths in the 2- to 10-mS range. These single-frame exposure times are short enough to freeze cardiac motion.

An optical system coupled the image intensifier to the cameras. Adjustable optical diaphragms were used to balance radiation dose and camera light levels. These diaphragms also allow service compensation for image intensifier degradation over time. The focal length of the optical system determines framing mode—the way in which the round output phosphor is represented on the rectangular cine frame (Fig. 2.8). Most laboratories used some form of overframing, in which the recorded field is less than the full active image intensifier area. In all such systems, it is essential to verify that only the recorded area of the patient is irradiated. In film-based systems, the ultimate quality of the recorded image depends nearly as much on the selection of cine film and its processing parameters as on the elements within the image chain.

Figure 2.8 Image framing. The focal length of the lens between the image intensifier and the video camera determines the visualized fraction of the output screen. Full overframing is shown on the left, partial in the center, and exact framing on the right. The choice determines the relative use of the image intensifier's physical field of view.

Video

Real-time fluoroscopic visualization is the enabling technology for invasive and interventional procedures. Video cameras and displays are the conduit between the image receptor and the observer's eye. Since the 1960s, this has been accomplished by placing a television camera in a position where it can (along with the 35-mm cine camera) view the output phosphor of the image intensifier. Solid-state charge-coupled device (CCD) television pickups have displaced analog video cameras in the last decade. The outputs of the video camera are converted into a digital television image and processed to enhance image quality before being stored or displayed. When the intent of imaging is simply to position a catheter or to perform a test injection, the cine camera need not operate, and the generator need provide only a low dose rate of radiation that is adequate to create a television image. Fluoroscopy thus involves <10% of the x-ray beam intensity that is used for permanent image recording (cineangiography). However, because fluoroscopic times are much longer than cine times, fluoroscopy typically provides more than half of the patient's total dose.

Older analog systems used the same interlaced scanning and display format as for broadcast television. Newer analog and all CCD systems use a progressive scanning and display format to allow higher video line rates (e.g., 1,024 versus 512 scan lines) and frame refresh rates. Therefore, video clips produced by such systems are not directly compatible with broadcast video components and recorders, although scan converters can be used to translate the video back to broadcast formats and standards.

Flat-Panel X-Ray Detectors

The image intensifier/video camera combination is currently being displaced by integrated digital image receptors (flat-panel detectors). Indirect detectors incorporate a charge-coupled device or photodiode visible light detector array in direct contact with the input phosphor. Direct detectors use a selenium layer to directly convert x-rays into an electron signal. Both designs generate a digital video signal with fewer intervening stages than described above for the phosphor image intensifier/video camera systems. Figure 2.9 schematically illustrates the structure of both flat-panel detectors.

The dose efficiency of a flat-panel detector is grossly similar to that of a modern image intensifier, so patient doses delivered by two fluoroscopes—one a flat-panel and the other an image intensifier—will be similar. However, flat-panel fluoroscopic systems often have a broader dynamic range and better dosimetric performance than older image intensifier–based systems owing to better dose management hardware and software in other parts of the fluoroscope.

The imaging behavior of a flat-panel system differs from an image intensifier/digital video system in one important respect. As shown in Fig. 2.10, when an image intensifier is

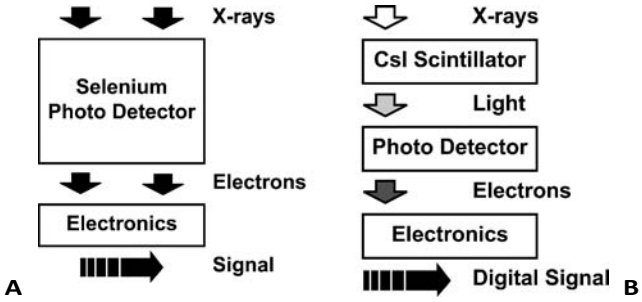

Figure 2.9 Indirect and direct digital (flat-panel) image receptors. The indirect (left) detector uses a CsI scintillator, virtually identical to that in an image intensifier, to convert the x-ray signal into light. A photodetector converts the light into an electron signal. This signal is then digitized. The direct detector (right) uses a selenium layer to directly convert the x-ray signal into an electrical charge distribution. This signal is then digitized.

zoomed, less and less of the patient is imaged by the tube's fixed-size output screen. Therefore each pixel in the zoomed image is smaller (relative to the patient) than for the unzoomed case; i.e., spatial resolution increases with zoom. In the flat-panel case, zooming simply uses fewer of the available pixels, so that the intrinsic spatial resolution does not increase with zoom. However, the digitally magnified image on the monitor may provide better detail coupling to the observer's eye, increasing the clinically effective resolution as a flat-panel system is zoomed.

Image Display and Processing

Digital images are processed before they are displayed (14). Image processing techniques include gray-scale transformations (changes contrast level), edge enhancement (improves the visibility of small high-contrast structures), smoothing (reduces the effect of noise in a single frame at the expense of image sharpness), and temporal averaging. This last function combines several image frames. It reduces noise while maintaining the sharpness of nonmoving structures. However, temporal averaging may blur moving objects. The type and extent of applied image processing can be configured by the service engineer and partially controlled by the operator.

Digital video facilitates functions such as fluoroscopic last-image-hold and instant replay of fluoroscopic and cinefluorographic images. Reviewing stored images instead of continuing fluoroscopy is an excellent means of patient and staff dose reduction.

DICOM PACS

Digital cardiac fluoroscopic and cinefluorographic images are typically produced using a nominal 1,024 × 1,024 pixel matrix. The bit depth can range from 8 to 12 bits (256 to 4,098 shades of gray). In the laboratory, these images are usually stored and displayed at full resolution. For archiving images, the 1995 DICOM standard

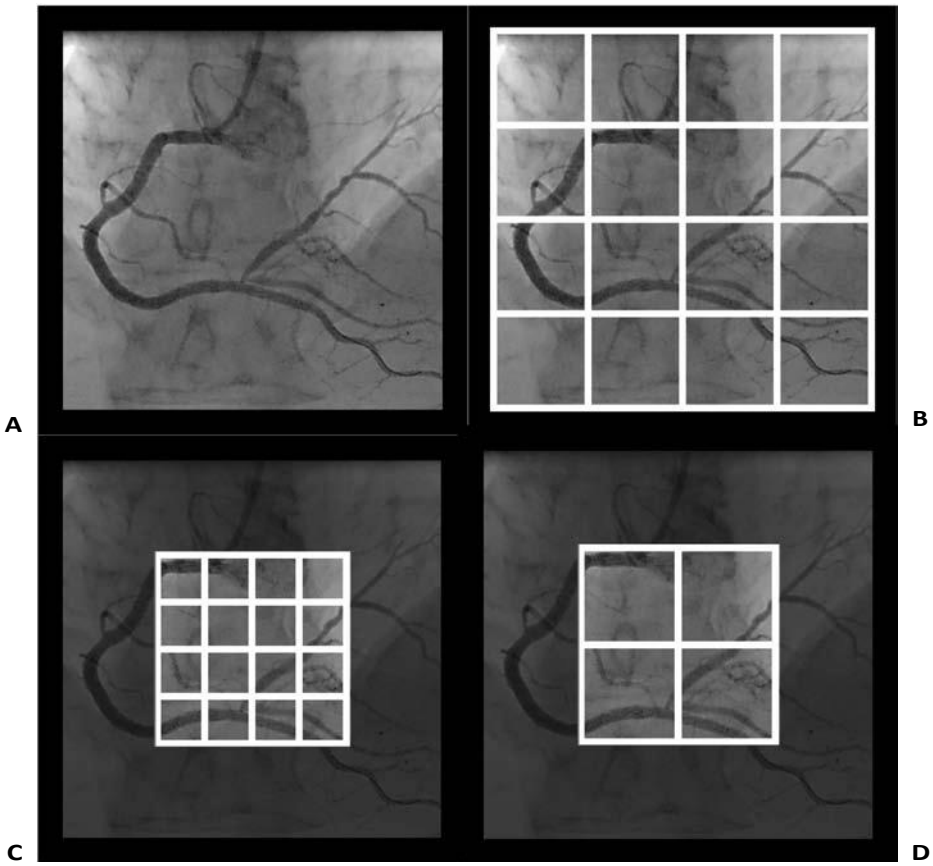

Figure 2.10 Zoom differences between image intensifiers and flat-panel detectors. The image before digitization **(A)**; full-field digitization for both systems—typically a matrix size of 1,024 × 1,024 **(B)**. When the image intensifier is zoomed **(C)**, the same matrix covers a smaller field of view; this reduces the effective pixel size. When a first-generation flat-panel is zoomed **(D)**, the pixel size remains the same; fewer pixels are used. Displays are usually electronically zoomed to fill the monitor. This does not increase physical resolution, but may improve the visibility of detail.

specified a 512 × 512 × 8 bit image format so that most studies would fit onto a single CD-ROM disc. The image standard has proved to be acceptable for most purposes over the decade since its release. Laboratory images are thus usually downscanned from their internal format into 512 × 512 × 8 bit before writing the study to a CD or transmitting it over a network for storage or remote viewing. Thus, image resolution when viewed in the laboratory is somewhat better than when the same cine run is recalled from storage. Higher-resolution images can be transmitted and stored at the expense of increased transmission time and storage space, and a general DICOM cine format maintains the same logical structure while permitting archiving of higher resolution images. Digital storage devices are available with an online capacity of tens of terabytes (TB). A 10-TB archive can store 20,000 to 30,000 cardiac cine studies in the 512 × 512 × 8 DICOM format. The cost of storage continues to decline by 10 to 30% per year, so online storage for all of a laboratory's archives is both technically and economically achievable.

DICOM images can be compressed (reduced in size) to save digital resources. Compression can be either reversible (the original image can be reconstructed exactly) or nonreversible. Reversible compression has been used in cardiology since the early 1990s, and the latest ACC document (15) specifically does not recommend the use of nonreversible compression for clinical decision making. Ongoing increases in device speeds, storage capacity, and network bandwidth have made the need for compression less urgent. It should be noted that common computer and Internet tools (MPEG, AVI) often lose much in compression, and they should be viewed with caution for diagnosis if DICOM images are available.

Image quality can be significantly degraded by poor viewing monitor performance. In DICOM terms, the world is divided into diagnostic workstations and review stations. The expectation is that primary medical decisions are made using images displayed only on diagnostic workstations. Facilities' quality programs are expected to include routine quality assurance of diagnostic workstations. Good practice dictates that a physician should have appropriate

confidence in the quality of any image display used for critical clinical decisions. Test procedures and images can be used to validate the performance of any imaging work- station.