# 21

## Fluoroscopy: Physics and Technology

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## 21.1 Introduction to Fluoroscopy

Fluoroscopy is a radiological examination that aims to provide visualization of dynamic processes as they occur. Fluoroscopy-guided procedures have become an important part of radiological practice since the beginning of the twentieth century. Fluoroscopy procedures use ionizing radiation for guidance as small instruments, such as catheters, guidewires, balloons, and stents, are manipulated through blood vessels or other pathways in the body. At first most of the procedures were diagnostic and had the scope only to identify clinical problems. Technologic advances over the past several decades led to a great increase in number and complexity of interventional procedures. As compared to open surgical procedures, interventional fluoroscopy procedures require a very small incision and permit shorter recovery times. They often have lower complication rates as well. As a result, these less invasive procedures have become very common, and are replacing open surgical procedures.

These procedures are used to treat a wide variety of diseases and disorders in virtually every organ system in the body. More
complex interventional fluoroscopy procedures are continually being introduced, such as endografts for the treatment of abdominal aortic aneurysms, vertebroplasty, kyphoplasty, and uterine fibroid embolization.

Modern fluoroscopes are used to determine a diagnosis from live display of patient anatomy, to position the patient for the recording of images and/or as a guidance for interventional procedure, depending on the clinical application. Such systems range from over- and under-table fluoroscopy systems to C-arm X-ray systems for surgery and angiography systems equipped for the diagnosis of vessel pathologies and minimally invasive treatment.

Fluoroscopy systems produce projection X-ray images (as in radiography). Therefore, fluoroscopy and radiography share some of the same imaging chain components, but differences exist. Fluoroscopy allows real-time X-ray viewing of the patient with high temporal resolution. Thus, it is necessary to update the image at rapid rates: typically 30 times/sec but, occasionally, up to 60 times/sec. In order to avoid radiation injury to the patient, low exposure rates are required. As a result, there is a significantly lower number of photons available to produce the fluoroscopic image compared with radiography. Therefore, the fluoroscopic image receptor must have specific characteristics to provide a visible image.

Image receptor technology has undergone impressive technological advancements in recent years. Before the late 1950s, fluoroscopy was performed in a darkened room with the radiologist viewing the light scintillations from a thick fluorescent screen (Patterson screen). This methodology is no longer permitted by international legislation and nowadays fluoroscopy systems are based on two different image receptor technologies: Image intensifier (II) and flat-panel detector (FPD). The image intensifier system coupled to digital video systems was developed first and it has been utilized for radiology imaging since the 1960s. Solid-state FPDs were introduced in dynamic imaging in 2001. They were originally designed for use in standard projection radiography. Solid-state digital radiography detectors provide on-line access to the electronic signal data, so the radiographic images are available to view in a matter of seconds after the exposure. Significantly, researchers found that with suitable technical optimization these solid-state detectors can be used equally well to record and read-out images at rates high enough to support fluoroscopy. FPD technology has led to larger rectangular field-of-view (FOV) systems with high spatial resolution and improved image quality, and has already replaced II detectors on high-dose interventional equipment.

The increase in number and complexity of interventional procedures led to a major concern for the growing radiation exposure of both patients and health personnel employed in interventional rooms. Those procedures in fact are often technically difficult and may require long fluoroscopy time and a large number of acquired images. Consequently, the doses received by both the patient and the personnel involved may be very high. This has raised highlighted dose-reduction tools and improvements in fluoroscopic systems have led to lower dose operation. Moreover, dose-monitoring technology has become a fundamental component of all interventional fluoroscopy systems.

21.2 Application Specific Design

Fluoroscopy has become important in several different clinical fields and today larger hospitals have several fluoroscopic suites dedicated to specific applications, such as gastroenterology, cystography, peripheral vascular and cardiac angiography, cardiac electrophysiology, and neurovascular imaging procedures. Consequently-fluoroscopic equipment has evolved over the years to become optimized for the clinical tasks for which they are intended and several different equipment configurations have been developed (see also Section II, Chapter 27 of this book).

The different configurations include radiography/fluoroscopy (R/F) tables, with either an under-table or over-table X-ray tube, and fixed and mobile C-arms. Each configuration includes the basic fluoroscopic imaging chain (see Section 21.3.1).

The R/F system is the most common fluoroscopic equipment configuration. It is used for a wide range of diagnostic and interventional procedures.

21.2.1 Fluoroscopic Equipment with Under-Table X-ray Tube

The X-ray tube and collimator are mounted below the tabletop with the image intensifier tower mounted above the table on a carriage that can be panned over the patient. This system can have a large tilting table that can be rotated from horizontal to vertical to put the patient in a head-down or head-up position for gastrointestinal (GI) procedures (e.g., barium enema), in order to utilize gravity to assist the movement of the contrast material through the esophagus, stomach, and bowel.

In addition to the standard fluoroscopic imaging chain, older R/F systems may contain a “spot film” device that allows placement of a radiographic cassette in front of the fluoroscopic image receptor, facilitating the acquisition of radiographs using the fluoroscopic X-ray source or with an overhead X-ray tube mounted on a ceiling crane in the room for regular radiography. A busy tray may be incorporated into the table for Computed Radiography (CR) (see also Section II, Chapter 12 of this book). The R/F system is the most common fluoroscopic equipment configuration. It is used for a wide range of diagnostic and interventional procedures.

21.2.2 Fluoroscopic Equipment with Over-Table X-ray Tube

A variation on this conventional fluoroscopy configuration is the remote controlled system shown in Figure 21.1, in which the X-ray tube and image receptor positions are reversed with the tube above the patient table and the image receptor below.

The system can be rotated to achieve necessary projections or to distribute contrast agents within a patient. It can also be configured vertically for seated examinations. All detector operations (e.g., pan, zoom, continuous vertical movement) and table movements can be fully controlled at an operator’s console featuring a joystick-type controller in a shielded control booth, protecting the staff from secondary radiation exposure.
In addition, a remote room typically has a motorized compression paddle that is used to remotely palpate the patient, and to manipulate air and contrast agent within the patient’s abdomen.

The clinical applications of this system are similar to those of under-table X-ray tube systems, including GI procedures as barium swallow and barium enema examinations.

21.2.3 Mobile C-Arms

Mobile fluoroscopes are fluoroscopes mounted on wheels that can be moved between locations. As seen in Figure 21.2, mobile C-arms offer a compact design, with smaller focus to image distances compared to other fluoroscopes, and they are provided with a mobile unit that contains the fluoroscopic controls and display monitors.

Most of these systems plug into a standard wall plug for power, and therefore are relatively low-power systems. High-power mobile systems (e.g., 80 kW) are also available. The image detector can be either an I1 or a FPD and is normally of small size (10–25 cm).

Mobile C-arms are useful when the expense of a permanent installation cannot be justified, or when imaging capability is needed briefly in several adjacent rooms, as for example, in the operating room. In fact, mobile C-arm units are commonly used for orthopedic and vascular surgical procedures, in addition to placement of devices such as pacemakers or feeding tubes. However, some mobile C-arm systems are also configured for angiographic and interventional procedures with high exposure rate output and digital subtraction angiography (DSA) imaging capabilities.

21.2.4 Fixed C-Arms

The fluoroscopic system is typically mounted on a C-arm apparatus. The X-ray tube and image receptor rotate contemporarily around a point called the isocenter that remains at the center of the FOV when the C-arm is rotated. These motions allow considerable flexibility in providing standard posteroanterior, lateral, and oblique projections.
Moreover, the fluoroscopy imaging chain can also be panned around the stationary patient and the table is floating in all the directions (vertically and horizontally) to allow moving the patient from side to side and from head to toe.

Because of its flexibility, the C-arm configuration has been incorporated into a number of different types of fluoroscopy systems. Figure 21.3 is an example of a fixed C-arm unit that is mounted from the ceiling. Floor-mounted models are also available.

Common applications for fixed C-arm units include peripheral, cardiac, neuroangiographic, and interventional procedures. A list of the most common procedures performed is reported in Table 21.1 (Miller et al. 2008).

Angiographic rooms are equipped with more powerful generators with high heat tube capacity and water or oil cooled X-ray tubes. Variable spectral shaping filters are often included to maximize iodine contrast while maintaining the patient dose at an acceptable level.

For peripheral angiography suites, a large detector is commonly used (30–44 cm), while image receptors used for cardiac imaging are generally smaller, owing to the small size of the heart. A typical size for a cardiac laboratory is 20–25 cm. The smaller detectors permit more tilt in the cranial caudal direction, as is typical in cardiac imaging. High frame rate (15–60 FPS) pulsed X-ray cineradiography operation is recommended to capture injected contrast medium as it moves through the vessels in the rapidly beating heart.

Systems designed for vascular/interventional radiology and cardiology/electrophysiology have sophisticated fluoroscopic capabilities, including variable frame rate, automatic beam filtration, and advanced dedicated image post-processing.

### Table 21.1

<table>
<thead>
<tr>
<th>Anatomical District</th>
<th>Procedure</th>
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<tbody>
<tr>
<td>Cardio</td>
<td>Coronary angiography (CA)</td>
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<td></td>
<td>Percutaneous transluminal cardiac angioplasty (PTCA)</td>
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<td></td>
<td>Stent and filter placement</td>
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<td></td>
<td>Radiofrequency cardiac catheter ablation</td>
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<td>Peripheral</td>
<td>Angiography</td>
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<td></td>
<td>Stent and filter placement</td>
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<tr>
<td></td>
<td>Percutaneous transluminal angioplasty (PTA)</td>
</tr>
<tr>
<td></td>
<td>Vascular embolization</td>
</tr>
<tr>
<td></td>
<td>Thrombolytic and fibrinolytic procedures</td>
</tr>
<tr>
<td></td>
<td>Percutaneous transhepatic cholangiography</td>
</tr>
<tr>
<td></td>
<td>Endoscopic retrograde cholangiopancreatography</td>
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<tr>
<td></td>
<td>Transjugular intrahepatic portosystemic shunt</td>
</tr>
<tr>
<td></td>
<td>Percutaneous nephrostomy</td>
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<tr>
<td></td>
<td>Biliary drainage or urinary/biliary stone removal</td>
</tr>
<tr>
<td>Neuro</td>
<td>Angiography</td>
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<td></td>
<td>Stent and filter placement</td>
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<td>PTA</td>
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<td>Vascular embolization</td>
</tr>
<tr>
<td></td>
<td>Thrombolytic and fibrinolytic procedures</td>
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Suites can be either single plane or biplane systems. Biplane systems use two C-arms (one ceiling mounted, the other floor-mounted—see Figure 21.4) that can be independently positioned around the patient for simultaneous digital acquisitions during a single contrast injection.

Biplane systems are used primarily for dedicated cardiac or neuroangiography and interventional procedures. By using two
separate imaging chains, it is possible to view frontal and lateral projections of the patient without introducing a delay while a single positioner is moved. In addition, biplane positioners allow recording in two projections during a single injection of iodinated contrast material. This is important because iodinated contrast is nephrotoxic, and the total volume of contrast that can be administered is limited by the patient body mass. This is particularly critical in pediatric catheterization laboratories, owing to the low body mass of pediatric patients, which severely limits the amount of contrast that can be administered during an imaging study, and the small size of blood vessels, which may require higher iodine concentrations for acceptable visualization.

21.3 Fluoroscopic Equipment

Fluoroscopic imaging systems use much of the same technology as radiographic systems, with some modifications and additions. Depending on the intended use, a fluoroscopic system may require a high-power generator and a high heat capacity X-ray tube. The major difference between radiographic and fluoroscopic equipment is the image receptor.

21.3.1 Imaging Chain

A standard fluoroscopic imaging chain is shown in Figure 21.5. The key components include an X-ray source (high-voltage generator and X-ray tube), spectral shaping filters, collimators (field restriction devices), an anti-scatter grid, an image receptor, an image processing computer, and a display device, and the hardware to allow the positioning of the X-ray source assembly and the image receptor assembly relative to the patient. Fixed fluoroscopic systems also have a patient-support device (Schuler 2000).

21.3.1.1 X-ray Source

The high-voltage generator and X-ray tube used in most fluoroscopy systems are similar in design and construction to tubes used for general radiographic applications. For special purpose rooms (e.g., high-dose procedures), extra heat capacity is needed to allow long fluoroscopic sequences and a high number acquisition runs acquired.

21.3.1.1.1 X-ray Generator

The X-ray generator produces electrical energy and allows the selection of kilovolt peak (kVp) and the tube current (mA) that is delivered to X-ray tube. The design of the generator (refer to Chapter 3 for more details) is similar to that of generators used for radiography and includes single phase and three-phase generators. Systems with both fluoroscopy and radiography modes normally use three-phase operation for radiography and single-phase operation for fluoroscopy.

Fluoroscopy includes either low continuous tube current and/or rapid pulsed exposure. Continuous fluoroscopy is still available in some fluoroscopic equipment, even if pulsed fluoroscopy has replaced it almost completely. For continuous fluoroscopy, the generator provides a fixed tube current while the fluoroscope is activated. For pulsed fluoroscopy, the exposure is delivered in short pulses of 3–10 msec. One advantage of pulsed fluoroscopy is the potential dose reduction compared to continuous fluoroscopy. But it is necessary to underline that the gain in dose depends not only on the pulse rate but also on the dose per pulse.

Another important advantage of pulsed fluoroscopy is that it provides better resolution since the motion blur occurring within each image is reduced because of the shorter acquisition time.

Modern systems are normally equipped with high frequency generators that can guarantee superior exposure reproducibility. Exposure reproducibility is critical especially for fluoroscopic
systems equipped with DSA, where differences in the tube voltage between mask and contrast images can cause incomplete subtraction.

21.3.1.1.2 X-ray Tube

The X-ray tube converts the electrical energy provided by the generator into an X-ray beam (see Chapter 2 for more details). The selection of X-ray tube characteristics depends on the specific clinical application. For radiography and fluoroscopy systems, bi-focus tubes are common. A small focal spot (0.6 mm) is used for fluoroscopy in order to minimize geometric unsharpness, and either the small or the large focal spot (1.0–1.2 mm) can be used for image recording when high tube currents are needed. A micro-focus (0.3 mm) can also be available when very high resolution is needed (e.g., interventional neuroradiology).

In modern X-ray tubes for angiography a flat emitter can replace the coiled filament. Flat emitters enable smaller quadratic focal spots that lead to improved visibility of small vessels reducing blur and increasing image sharpness in all directions. Moreover, pulsed fluoroscopy can be accomplished by using a grid-controlled or grid-switched X-ray tube instead of operating the generator in pulsed mode. A grid-controlled X-ray tube uses a negatively biased grid near the filament to stop the flow of electrons from the cathode to the anode, preventing unwanted X-ray production between radiation pulses. The output is controlled within the X-ray tube itself and the unnecessary soft radiation emitted by ramping can be eliminated. The grid control pulsing allows the parameters (kV, mA, and ms) to be adjusted within the duration of a single pulse.

The combination of grid pulse and flat emitter technology potentially produces a sharp temporal pulse profile removing the additional dose due to the high-voltage cable discharge. As a consequence both the dose and the blur are reduced.

Another crucial characteristic of X-ray tubes used in angiography and interventional procedures is the heat capacity. Because of the rapid image recording requirements, heat can build up quickly. To improve heat dissipation, high-speed rotational anodes are used, with frequencies up to 200 Hz. In addition, a circulating water or oil heat exchanger with cooling fans is commonly installed.

For an X-ray tube in a dedicated angiography or interventional system, maximum FOV size requirements may limit the heat capacity or minimum focal spot size usable. When a large FOV is needed, the anode angle must be large enough to allow adequate coverage without cutoff of the beam intensity. However, for the same effective focal spot size, a large anode angle results in a reduced focal spot width, which reduces the rate of heat dissipation lowering the heat capacity of the tube.

21.3.1.2 Filtration

Fluoroscopic and angiographic equipment are provided with different types of filters used to optimize both the spectrum and the shape of the X-ray beam.

21.3.1.2.1 Spectral Filtration

Filtration material is added to attenuate low-energy X rays from the beam. Low-energy X rays are absorbed in patient tissue without being transmitted to the image receptor, contributing to patient dose with little improvement in image quality. The added filtration in fluoroscopy systems has increased in recent years, as pulsed systems with high mA capabilities started replacing older continuous fluoroscopy systems.

Aluminum is the most common added filtration material. Copper can also be used for low kV beams used to match the barium or iodine k-edge, reducing the dose while maintaining the image contrast for angiographic applications.
The new generation of X-ray generators provides higher peak power and allows using high filtration while maintaining the image quality (e.g., noise level) at a reduced dose. The higher peak power also allows reduction of the exposure time per pulse thus reducing the motion artefacts.

Insertion of this added filtration in the beam path may be user-selectable, providing the operator with the flexibility to switch between low-dose and higher-dose modes as conditions dictate during a fluoroscopic procedure. In other systems (like the angiographic and interventional systems), the added filtration is automatic, based on beam attenuation conditions, to achieve a desired level of image quality and dose savings.

21.3.1.2 Collimators

The collimator contains multiple sets of radiopaque shutter blades that define the shape of the X-ray beam. Two sets of blades are generally present within the collimator: round and rectangular, matching the shape of the image receptor. The collimator conforms the X-ray beam to the FOVs. The collimators on fluoroscopy systems are motorized, which allows the collimated X-ray beam to change size in response to adjustment of the source-to-image distance (SID) or when the FOV is electronically changed, in order to limit the X-ray beam to no larger than the FOV. However, further manual collimation might be beneficial for the operator to further collimate the beam to the area of clinical interest thus limiting the exposed volume of tissue, which results in less scatter production and better image contrast.

21.3.1.3 Equalization Filters

In addition to added filtration and beam collimation, some fluoroscopy systems also have attenuating equalization filters, alternatively named wedges filters, as means to provide further beam shaping in addition to collimation. The filters are made from tapered lead–rubber or lead–acrylic sheet and are partially transparent to the X-ray beam. The position of the wedge filters is selected by the operator to provide additional attenuation at specified locations in the X-ray field (such as the pulmonary space between the heart and chest wall) and reduce excessive brightness in some regions in the image. The use of wedges can thus reduce the glare from these areas, equalize the X-ray beam incident on the image receptor, and lower radiation dose to the patient, improving the operation of the automatic exposure control (AEC) system. The edges of the blades may be straight or shaped to conform to anatomic parts.

21.3.1.4 Imaging Detector

The basic product of a fluoroscopic imaging system is a projection X-ray image. However fluoroscopy produces an extremely high number of images with high temporal resolution. As an example, the average fluoroscopy time for a coronary intervention is about 10 min and images are acquired at 15 per second, corresponding to a total number of 9000 individual images. Due to the extremely large number of images, for radiation dose reasons, fluoroscopic systems must produce each image at a dose of the X-ray much lower than the dose used in radiography. Consequently, a very sensitive low-noise detector is needed. IIs and FPDs, the two imaging detectors used for the fluoroscopic equipment, operate in a mode that is several thousand times more sensitive than a standard radiographic detector and, in principle, can produce images using several thousand times less radiation (standard fluoroscopy uses a dose to the detector of few nGys per image, whereas a computed radiography (CR) system requires a few mGys per image).

21.3.1.3.1 Image Intensifier-Based Systems

A detailed scheme of modern image intensifier system is shown in Figure 21.6 and includes the image intensifier and the optical system.

21.3.1.3.1.1 Image Intensifiers

The image intensifier converts X-rays into visible light suitable for being captured by a video camera and displayed on a video display monitor. At the same time, it amplifies the image brightness by about 10,000 times. Major components are contained within an evacuated bottle (Figure 21.7) and include:

- **Input layer to convert X-rays to electrons**: The input layer is, in turn, made of different components. The X-rays strike the input window that is made of a curved, thin layer of metal or glass, which prevent the air entering the II. Then they pass through the 0.5-mm-thick aluminum substrate layer, which supports the input phosphor and the photocathode layers. After passing through the Al window, X-rays strike the input phosphor layer that is made of cesium iodide (CsI). The input phosphor absorbs the X-rays and converts them into light photons. The input phosphor is deposited as long, needlelike crystals to channel the light photons to

![FIGURE 21.6 Schematic representation of an II based system.](image)
the next component layer with minimal lateral spreading to reduce blur. The light photons emitted from the input phosphor are then absorbed in the photocathode, which is a thin layer of antimony and alkali metals, that is placed in close proximity to the input phosphor and converts the light photons into electrons.

IIIs are available with different diameter input windows of 10–40 cm. The selection of the diameter depends on the maximum FOV requirements of the clinical application. Fluoroscopic systems designed for extremities may be configured with a 10–15-cm-diameter image intensifier, whereas a 40-cm-diameter unit is useful for imaging the abdomen or peripheral vasculature.

- **Electron optics to focus the electrons**: The electrons emerging from the photocathode are focused, accelerated, and multiplied in number through the vacuum by the electron optic system. This system consists of three charged electrodes and an anode plate at the output layer. These components create a large electric field that accelerates the electrons. The electrodes in the chain focus the electrons towards the anode.

  If the voltage of each electrostatic plate is not adjusted correctly, the electrons will not pass through the appropriate focal point of the image intensifier, causing the image to be blurry with a loss of spatial resolution; this blurriness is referred to as a defocusing effect.

  The kinetic energy of each electron is dramatically increased by acceleration due to the voltage difference between the cathode and anode, resulting in electronic gain, which is the first source of intensification. The other source of intensification is minification gain. Minification gain is a result of the reduction of a large X-ray image at the input phosphor onto the smaller diameter output phosphor.

  As a result of the acceleration of the electrons and image minification, the illumination level of the output image compared with that of the input image is greatly increased. This brightness gain ranges from 5000 to 20,000.

  The conversion factor (namely ratio of a measure of the luminance at the output phosphor and the X-ray Incident Air Kerma Rate (IAKR) at the input phosphor) is a measure of the gain of an X-ray II (XRII) and is most commonly used for the specification of XRII performance. The conversion factor decreases over time and may ultimately fall to a level that requires replacement of the XRII.

- **Output phosphor layer to convert the electrons into a visible image**: The output phosphor consists of a thin (4–8 μm) powder phosphor, typically zinc cadmium sulfide doped with silver (ZnCdS:Ag – P-20). The output phosphor converts the electrons back into visible light photons at a frequency of about 530 nm (green spectrum). These photons are then transmitted out of the image intensifier through a glass output window. The output window is part of the vacuum enclosure and must be transparent to the emission of light from the output phosphor. The output phosphor is coated directly onto the output window.

    Some fraction of the light emitted by the output phosphor is reflected inside the glass window. Such stray light reflecting inside the output window contributes to the veiling glare, which can reduce image contrast. Using a thick clear glass window coated inside with a black pigment to absorb the scattered light, the side of which is eventually struck by internally reflected light, reduces this veiling glare.

    Projecting the image with a curved input phosphor (necessary for proper electron focusing) to the flat output phosphor creates a distortion known as “pincushion.” Pincushion distortion results in a stretching of the physical dimensions in the periphery of the image (see Figure 21.8). Therefore, for improved accuracy, it is best to position the desired anatomy in the central area of the FOV.

    Moreover, the longer paths covered by the peripheral electrons reduce the concentration of electrons that impact on the side of the output phosphor and cause the phenomenon called vignetting, with the center of the image brighter than the edges.
21.3.1.3.1.2 Optical System The visible photons exiting the output window reach a video camera optically coupled to the phosphor screen through an adjustable aperture and lens. The video signal is then displayed directly (or digitized), post-processed in a computer, and rendered for display. The television system allows for real-time viewing of the fluoroscopic image by several people at once from one monitor or multiple monitors. In addition, fluoroscopic units can be equipped with an analog-to-digital converter to digitize the video camera voltage signal for additional processing and image recording.

- **Optical Coupling**: A light-sensitive camera is optically coupled to the output screen of the II and is used to relay the output image to a video monitor for viewing by the operator. The optical coupling system consists basically of three elements: (i) a collimating lens to shape the divergent light from the output phosphor into an almost parallel beam, (ii) an aperture to limit the amount of light reaching a video (or TV) camera, and (iii) a lens to focus the image on to the video camera.

  The aperture can either be fixed or variable, the latter usually being under automatic control. The amount of light that reaches the camera can be controlled varying the size of the aperture, adjusting the quality of the fluorescent image and the X-ray exposure rate. When the hole is set to a small size, much of the light from the output window is blocked from reaching the video camera. As a result, the radiation exposure has to be increased to maintain the light level at the camera, producing a fluoroscopic image with low noise. Alternatively, when the aperture is set wide open, the radiation exposure level is low and more image noise is apparent.

  The optical distributor may include a partially silivered, beam-splitting mirror, which directs a portion of the light from the image intensifier output window to an accessory device for image recording and passes the remainder to the video camera. A circular aperture is also included to set the proper light level required by the video camera.

- **Video Camera**: The video cameras used in XRII systems were originally vidicon or plumbicon analog devices borrowed from the broadcast television industry. A basic video camera consists of a vacuum tube cylinder (approximately 2.5 cm in diameter) with a photoconductive target and a scanning electron beam. The optical coupling lens focuses the image intensifier output image onto the target, forming a latent charge image from the charge carriers within the photoconductive layer. The electron beam, which scans across the target in a raster scan pattern, reads out this latent image. As the scanning electron beam moves across the target, a current signal, with an intensity modulated by the charge image present on the target, is produced. This signal represents the two-dimensional image as a continuous series of raster lines with varying voltage levels. The video signal, which is represented as voltage versus time, is transferred by wire to the video monitor.

In later systems, digital cameras based on charge-coupled device (CCD) image sensors or complementary metal oxide semiconductor (CMOS) technology came into common use.

CCD cameras consist of a solid-state array of discrete photoconducting sensors. Optical light from the output phosphor is converted to electrons in the amorphous silicon photoconducting layer of the CCD and stored as pixels until it is read out as voltage pulses representing the two-dimensional image. Stored charge that has accumulated during an exposure is read out using parallel and serial shift registers that move charge from column to column and row to row in a "bucket brigade" fashion, creating an analogue signal. The signals from the column amplifiers are synchronized and convert each row of accumulated charge into a corresponding digital value in the digital frame buffer.

The CMOS device has independent detector elements, each comprised of a storage capacitor and transistor switch.

In either output (CCD or CMOS) a digital projection image is produced, and then converted to a video signal that is used to drive one or more monitors in the fluoroscopy suite.

Compared with traditional video cameras, solid-state cameras are preferred because they are smaller, require less power, and have a longer lifetime. Moreover they guarantee a wider linear dynamic range. Incorporating drains in each cell that direct excess charge to reduce or eliminate image blooming (image distortion caused when the input signal exceeds the dynamic range of the video camera). This modification is at the expense of the fill factor though, and reduces the overall quantum detection efficiency (QDE) of the camera.

- **Monitor**: The voltage signal produced by the video camera is converted into a visible image by the monitor. The monitor consists of a vacuum chamber with a phosphor screen and scanning electron beam. The electron beam moves across the phosphor screen in horizontal raster scan lines with the intensity variation controlled by the camera voltage signal, thus reproducing the image from the video camera target.

- **Image Recording**: A fluoroscopic imaging system may include additional devices to record images during an examination. Recording methods include spot film devices, film changers, photospot cameras, cine cameras, and digital photospots. The selection of the optimum recording method for a particular clinical application depends on the operational characteristics of the device and image quality requirements of the examination.

Spot film devices are used to acquire a radiographic image with a screen-film cassette. When a spot film is desired, a button is pressed and the radiographic cassette is extended in front of the XRII, behind an
anti-scatter grid. After the cassette is exposed, it is ejected and manually exchanged for an unexposed cassette, which is retracted into the lead-shielded enclosure until needed. Collimation can be automatically varied to produce multiple image formats. The image quality is the same as a radiographic film and large-film image recording is possible. Clinical applications of spot film devices include GI imaging, genitourinary imaging, arthrography, and myelography.

Photospot cameras are mounted on the optical distributor accessory device port to record images rapidly during the fluoroscopic examination and record the image intensifier output on rolled or cut film to produce images about 10 cm in diameter. Photofluorography is generally used for the same clinical examinations as spot film devices. A cine camera may also be mounted as an accessory image recording device to acquire images on 35-mm film.

Cinefluorography is typically used for cardiac catheterization procedures to record rapid rate images of the beating heart. In newer fluoroscopic systems, these film recording methods are replaced with digital image recording.

Digital photospots are acquired by recording a digitized video signal and storing it in computer memory. The image quality can be enhanced by the application of various image processing techniques, including window-level, frame averaging, and edge enhancement. However, the spatial resolution of digital photospots is less than that of film images.

21.3.1.3.2 Flat Panel Detector Based Systems

In recent years FPDs gradually replaced XRII and video camera components of fluoroscopic systems (Figure 21.9).

When flat panel X-ray detectors have been introduced in radiography, they offered the advantages of a “digital camera” compared with existing technologies (Antonuk et al. 1995).

In fluoroscopic applications, the challenge for FPDs has been the requirement of low dose per image frame, meaning that the inherent electronic noise of the detector must be extremely low and the required dynamic range high.

FPDs consist of an array of individual detector elements. The elements are square, 140–200 microns per side. The size of the entire array ranges from 20 × 20 cm–40 × 40 cm. However, some manufacturers specify the size of the FPD by providing a diagonal measurement, and others quote the edge dimension. A FPD may contain 1.5–5.0 million individual detector elements and it is a challenge to manufacture a uniform array with few defective or degraded elements.

Both indirect and direct X-ray conversion modes are used with thin film transistor (TFT) panels for fluoroscopy applications even though the indirect conversions based systems represent the majority (Korner et al. 2007).

In indirect solid-state systems, the X-ray energy is first converted to light and then to an electronic signal. An individual detector element consists of a scintillation layer, which usually is composed of thallium-activated CsI which attenuates the incident X-rays and produces light. As for the scintillator used in II systems, the CsI is deposited in needle-like crystals, which drive direct light toward the photodiode located below with reduced lateral scatter. The amount of light produced is directly related to the amount of incident X-ray flux. Each detector element has a transistor and a capacitor, in addition to a photodiode, which converts the X-ray induced light from the phosphor into a corresponding charge. A schematic representation of an indirect FPD is reported in Figure 21.10.

For direct detection fluoroscopy detectors, the selenium is used as semiconductor to produce X-ray induced charge directly. The charge is collected under a voltage to ensure that the signal is captured within the same detector element as the X-ray absorption event.

In both types of systems, each detector element has a capacitor, which accumulates and stores the signal as an electrical charge proportional to the incident X-ray flux, and a TFT, which serves as a switch (Schiebel et al. 1994). A signal is sent to all the TFTs in a row of detector elements and the signals from the capacitors in the elements of that row are simultaneously discharged and the charge is sent to a charge amplifier. The discharged capacitors

![FIGURE 21.9 Schematic representation of FPD-based system.](image-url)

![FIGURE 21.10 Schematic representation of indirect flat-panel detector.](image-url)
are thus ready for acquiring the next frame. The same mechanism is performed row by row until the full array is read. Even in continuous fluoroscopic acquisitions, the charge continues to be stored in each capacitor in a steady-state situation so that X-ray information is always collected as the flat-panel image receptor provides real-time fluoroscopic presentation of image data.

FPDs have a many advantages compared to IIs. They replace the IIs, optics, video camera, digital spot film device, and cine camera and, thus, they come in a much lighter and smaller package, which allows more flexibility in movements and patient positioning. FPDs show a higher dynamic range, as shown in Figure 21.11.

Moreover FPDs do not suffer the previously described distortion effects of the IIs, for example, defocusing, pincushion distortion, veiling glare, or the vignetting. Another advantage is that the spatial resolution depends on the dimension of the detector element and it is independent of the selected FOV. Unlike II systems, which progressively use more radiation as the FOV decreases, FPDs do not require the entrance kerma rate to increase as the FOV is changed. Nevertheless, the radiation dose to the image receptor is increased by a factor of approximately \(1/\text{FOV}\), because with smaller FOVs, magnification of the surface area makes the image noise more apparent to the observer. In FPD systems, increased radiation for smaller FOVs is used to reduce the optical perception of noise. However, this increase in radiation is substantially less than that used with image intensifier systems, which varies approximately with \(1/(\text{FOV})^2\).

However, flat panel image receptors suffer from additive noise sources, including read noise and, therefore, perform poorly compared with XRIIs at low entrance kerma rates. Moreover part of the incident radiation is actually hitting the electronic elements and it is not used to form the image. Consequently the detective quantum efficiency (DQE) of the FPD at very low doses is lower than the DQE of an II system.

To compensate for this, binning (i.e., grouping the responses of four detector elements together) is often applied. This will reduce the entrance dose rate to 25% of the ungrouped rate. Binning could reduce the image noise but it negatively affects the spatial resolution because the effective area of each image pixel is four times larger.

In recent years, advances in CMOS imaging device technology have led the way to adoption of CMOS technology in medical imaging. The basic detection principle for such devices is the same as described above for CsI/a-Si-based active matrix integrating detectors. However, a CMOS sensor replaces the a-Si photodiode and readout structure. The electrical properties of crystalline Si allow realization of high performance analog or digital circuitry. This, for instance, makes on-pixel amplification possible, reducing readout electronic noise, which helps to move towards the goal of quantum-noise-limited X-ray detectors at very low doses (few nGy).

### 21.3.1.4 Grid

The focused anti-scatter grid serves the same purpose in fluoroscopic imaging as it does in radiographic imaging, explicitly, removing contrast degrading scattered radiation from the X-ray beam. However, use of grids requires an increase in radiation exposure. The grid ratios for fluoroscopy range from 6:1 to 10:1, which is generally lower than common radiographic grid ratios (8:1–16:1). For fluoroscopy, removal of the grid may be desirable to reduce patient dose when the amount of scatter produced is low. Image contrast loss will be minimal when the FOV is reduced or the patient or body part examined is small, as in pediatric imaging.

### 21.3.1.5 Patient Support

The patient table and pad must balance adequate strength to support the patient’s body weight while minimizing X-ray attenuation. This can be accomplished with carbon fiber composite materials and thin foam pads.

### 21.3.1.6 Image Display

Fluoroscopy requires high-quality video displays that allow users to appreciate fine details and subtle contrast differences in the anatomy of interest. Modern systems feature high resolution flat-panel LCDs with high maximum luminance and high-contrast ratios. These displays should be calibrated to a standard luminance response function to ensure that the widest range of gray levels is visible.

The newest interventional/angiographic systems feature 60° diagonal high-definition displays supporting up to 24 different video input sources that can be arranged in various ways on the single large display monitor. Display layouts can be uniquely customized and saved for individual physician preference.

### 21.3.1.7 Automatic Exposure Control

Fluoroscopic X-ray systems make use of a set of rules that control the system’s response to dynamic changes in imaging conditions (e.g., patient thickness and attenuation, focus to detector distance, projection angle, and so on) changing the exposure rates incident on the imaging detector automatically. The scope is to maintain the absorbed energy fluence per pixel used for each frame at a constant level in order to keep the signal-to-noise ratio (SNR) in the image approximately constant.

The control of generator exposure factors is called automatic dose rate control (ADRC) or automatic exposure rate control (AERC). In the past, this was referred to as automatic brightness control (ABC) since the aim was to keep the overall image brightness on the monitor at a constant level.
Fluoroscopic AEC may use the signal from a sensor such as a photodiode or a photomultiplier tube or, more commonly, the signal from the video camera or directly from a flat panel image receptor, to determine necessary adjustments of fluoroscopic technique factors such as tube voltage and tube current.

The number of factors controlled depends on the type of the equipment. Old II systems have fixed beam filtration, continuous mode, fixed optical aperture, and limited image processing. The only parameters that can be controlled to maintain the signal were the kV and mA, which were modified according to the anti-isowatt curves (the kV and mA increased or decreased simultaneously). Systems are designed to ensure that the design limits of the X-ray generator and the heat rating of the X-ray tube are not exceeded. In addition, the allowed combinations of generator parameters are such that radiation exposure rates do not exceed the regulatory limits.

For a given curve, when the kVp and mA reaches the limit of X-ray tube loading, the curve automatically switches to isowatt control. Along the isowatt line, the kVp increases while the mA decreases to maintain the product of kVp and mA constant.

The selection of fluoroscopic technique factors follows predetermined curves that are stored in the generator and which usually allow for some choices, including a standard curve, low-dose curve, and high-contrast curve (Figure 21.12).

The complexity of fluoroscopic AEC increases in modern systems where the AERC controls contemporarily also additional parameters such as pulse length, added filtration, and variable aperture setting in order to maintain or improve image quality while reducing the patient dose.

An example of parameters (and limitations) that the ADRC system takes into account is reported in Table 21.2.

According to the report of AAPM TG125, two general approaches of spectral beam filtration are implemented: the “traditional” method and the “program-switched” method.

The traditional method refers to a fixed factory-installed filter, which does not change with the program selected. This approach is still used in mobile C-arm systems.

In the anatomical program-switched method predetermined combinations (different materials and/or thickness) of added spectral filtration can be automatically switched under program control. The filter selected can also be linked to one of the fluoroscopy exposure rate control settings. Once a fluoroscopy mode is selected, the spectral filtration is set and does not depend on the variations in patient attenuation. In addition, the system could be programmed to automatically change the filtration with change in source-to-image distance (SID).

The most advanced method is the so-called Seissl method, where filters are changed in a dynamic way. The exact scheme for determining which filters are selected is proprietary. Figure 21.13 represents the behavior of an AERC system based on the Seissl method.

Lin et al. investigated the behavior of such methods; refer to their publications for more details.

As an adjunct to the ABC/AERC an automatic gain correction (AGC) of the detector is used to compensate for changes in the signal strength among the different exposure settings (e.g., fluoroscopy, acquisition, DSA). Image “gain” is used to adjust the digital pixel values and hence the appearance of an X-ray image. When the requested detector output is not reached, the gain control makes the image appear, in terms of average luminance, as though the detector output was not actually too low. This is operated automatically, utilizing the difference between the measured and requested detector output.

In addition, in II systems, the size of the aperture in the optical system transmitting the light from the output phosphor to the video camera can also be utilized to maintain a constant video signal level. In most cases a combination of ABC/AERC control logic, AGC and optical aperture modulation will be employed as a means of maintaining a reasonably constant video signal.

The source-to-image (SID) output compensation is also implemented as means of keeping the maximum radiation output in accordance with the regulatory maximum exposure rate requirements. The SID output compensation consists in the adjustment of the maximum radiation output in accordance with the X-ray tube load limits and the regulatory maximum exposure rate requirements.

### 21.3.2 Operating Modes

#### 21.3.2.1 Fluoroscopy

Continuous fluoroscopy has been the most basic form of fluoroscopic imaging for long time. The X-ray beam is on constantly and a video refresh rate of 25 or 30 frames/s yields a frame integration time of 40 or 33 ms.

![FIGURE 21.12](Typical basic AEC control curves for different tasks.)

### Table 21.2

<table>
<thead>
<tr>
<th>Parameters Used by AEC Systems to Optimize Exposure</th>
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<tr>
<td><strong>Control Parameters</strong></td>
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<tr>
<td>Peak voltage</td>
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<tr>
<td>Spectral filtration</td>
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<tr>
<td>Tube current</td>
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<tr>
<td>Pulse width</td>
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<tr>
<td>Detector dose</td>
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<td>Frame rate</td>
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<table>
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<th>Constraints</th>
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<tr>
<td>Tube/generator</td>
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As previously introduced, most modern fluoroscopic equipment is capable of operating in pulsed fluoroscopy mode. When configured properly, pulsed mode offers several advantages over continuous mode, including lower radiation dose. Moreover, in pulsed fluoroscopy, there is also an improvement in image quality in terms of motion blur reduction due to the lower integration time and the reduced tube loading at low pulse rates. While pulsed fluoroscopy produces sharper images, the reduction in temporal resolution at low frame rates may be unacceptable for rapidly moving organs or instruments within the body. Higher frame rates provide superior temporal resolution for these cases. Moreover, since the temporal response of the human visual system has a typical integration time of approximately 0.1 s, it has the capacity to integrate several frames of pulsed fluoroscopy during a single integration cycle. Consequently, fluoroscopic images appear noisier as the pulse rate decreases for the same IAKR per frame. When changing from one pulse rate to another, the input air kerma per pulse can be adjusted to account for this phenomenon.

Most of the fluoroscopic procedures do not require a high frame rate. For example, in peripheral angiography a frame rate of 4 FPS can be used.

Pulsed fluoroscopy at selectable frame rates (typically 30, 15, 7.5, and 3.75 FPS) allows to reduce temporal resolution when it is not needed, sparing dose.

### 21.3.2.2 Digital Acquisition Imaging

Digital acquisition imaging refers to a mode of operation in which high quality images are recorded and stored for analysis. IAKRs, and therefore patient dose rates, are at least one order of magnitude higher in digital acquisition mode than in the fluoroscopic mode. The detector gain is adjusted accordingly.

Digital acquisition images can be acquired at frame rates ranging from 1 to 30 FPS, or as individual images, which are often referred to as single shot images.

### 21.3.2.3 Digital Subtraction Angiography

DSA is a technique in which sequential images acquired after the injection of a contrast are subtracted from a mask image that includes only the anatomical background (acquired before the injection). This subtraction reduces anatomical noise and increases the contrast of the blood vessels in the subtracted images. Both the mask and fill images undergo a log transform before subtraction. The final result is an image in which the signal in the contrast-filled vessels depends only on the amount of contrast in the vessel, and not on the background.

As quantum noise sums in quadrature when images are combined, the noise level in the subtracted image is higher than the noise level in the constituent images. This increase in noise implies that DSA will require higher exposures than digital acquisition imaging if similar image noise levels have to be maintained. However, the reduction in anatomical noise achieved with DSA may offset part or all of the increase in image noise, and advanced techniques such as mask averaging can be used to reduce the exposure requirements for DSA imaging.

The major issue with DSA imaging is artefacts, which are especially due to patient motion between the capture of the mask and fill images. These types of artefact can be reduced retrospectively in some cases through the use of processing techniques such as manual or automatic pixel shifting of the mask image, or re-masking through the selection of a different mask frame for subtraction.

### 21.3.2.4 Roadmapping

Roadmapping is an imaging mode used to create a map of vascular anatomy that aids the navigation of catheters within tortuous vessels. A roadmap can be generated very simply by using a stored image of a contrast-filled vessel, or in a more complex fashion by using the peak opacification in each image pixel obtained from a series of post-injection images. This is essentially a maximum intensity projection image of the contrast-filled vessel.
vessel and ensures a relatively uniform signal throughout the vessel, as it is less affected by contrast washout.

The acquired image can be shown on the monitor next to the live fluoroscopy monitor. In this way, the path of the vessel can be seen on one monitor, while the angiographer advances the catheter viewing in real-time on the other monitor. Another approach to road mapping is to use an injected or subtracted image as an overlay onto the live fluoroscopy monitor. In this way, the angiographer has the vascular "road map" superimposed on the fluoroscopy image and can orient the guidewire or catheter tip to negotiate the patient's vascular anatomy.

21.3.2.5 Rotational Angiography

Advances in angiographic interventions, including vascular stent and stent graft placement, thrombolysis, transcatheter embolization, and targeted intravascular oncologic procedures, have increased the need for accurate 3D characterization of vessels and adjacent structures. In addition, non-angiographic procedures, as percutaneous drain and stent placement and radiofrequency ablation frequently involve complex, anatomical relationships, which are difficult to characterize with 2D fluoroscopic images. For this purpose, sets of CT images are often acquired before and after the interventions. These considerations lead to efforts to develop systems for generating 3D dataset using the same technology used to acquire fluoroscopic images.

The first 3D method introduced was the rotational 3D-DSA, which allows 3D renderings of digitally subtracted contrast-enhanced vessels. With this technology, multiple DSA images are generated at various projection angles by rotating a conventional angiography unit around the patient. 3D image sets are derived using a cone beam back projection reconstruction algorithm.

Shortly after the 3D-DSA, 3D digital angiography was developed, which uses un-subtracted digital images and allow the 3D visualization of high-contrast structures, including bones and contrast-enhanced vessels. Three-dimensional digital angiography has the potential advantage over 3D DSA of no misregistration artefacts and lower patient dose. However, the detectability of low-contrast structures is limited.

In the late 1990s, experimental systems were developed using cone beam computed tomography (CBCT). CBCT enables the generation of an entire volumetric dataset in a single gantry rotation by using a 2D detector system. CBCT mounted on a C-arm was initially performed using an II system. However, II systems coupled with CCD suffer from limited spatial resolution and were soon replaced by the FPD and today all C-arm CBCT units employ FPDs.

C-arm CBCT allows the acquisition of a complete volumetric data dataset covering a large anatomic region in a single rotation of about 200° (180° plus fan angle) of the X-ray source and detector in a single orbit about the patient. C-arm CBCT systems currently available require 5-20 seconds for image acquisition.

The reconstruction algorithm is a modification of the algorithm initially described by Feldkamp. The images can be reconstructed in axial, coronal, and sagittal planes, or in arbitrary curved planes. The fidelity of the images is not as good as with whole body CT scanners, because the rotational axis on the C-arm system is not as rigid.

Moreover, while the high efficiency of a2D detector allows excellent low-contrast detectability and potentially higher spatial resolution, there is a significant increase in scatter radiation with C-arm CBCT due to wider X-ray beam collimation that limits the dimension of the voxel and leads to a significant degradation of image quality (e.g., image artifacts, decreased contrast-to-noise (CNR), and inaccuracies in CT number calculations).

To account for the increased scatter, multiple anti-scatter techniques have been investigated for use with CBCT systems including, anti-scatter grids, software correction algorithms, beam-stop scatter map ping, and adjustment of object-to-detector distance. The most effective is limiting the dimension of the field-of-view, which minimize both the scatter-to-primary ratio and patient dose.

The applications of C-arm CBCT imaging are many. In interventional neuroradiology, CBCT is particularly useful in intra- and extra-cranial arteriography, for aneurysm characterization. Recent investigations suggest that current generation C-arm CBCT systems should also be able to reliably detect intracranial hemorrhage. C-arm CBCT has also demonstrated utility in the repair of endoleak following endovascular repair of abdominal aortic aneurysms. Moreover, studies using C-arm CBCT for transjugular intrahepatic portosystemic shunt placement and transcatheter arterial embolization are encouraging.

21.3.3 Image Performances, Tools and Advanced Post-processing

As for the other X-ray imaging modalities, the factors that must be taken into account when considering image quality in fluoroscopic imaging include contrast resolution, spatial resolution, and temporal resolution (Cowen et al. 2008). While each of these quantities is influenced and limited by the design of the fluoroscopic equipment, they are also highly dependent on equipment configuration and use.

21.3.3.1 Contrast Resolution

Subject contrast is inherently poor in fluoroscopic imaging compared to radiography, especially at the high kV values used to maintain patient dose at an acceptable level. Fluoroscopic systems with different dose settings (selectable at the console) allow the user flexibility from patient to patient to adjust the compromise between contrast resolution and patient dose. Contrast is greatly improved through the use of radiopaque markers on catheters and other instruments, and through the use of exogenous contrast agents. Contrast agents for fluoroscopy are selected on the basis of their chemical properties, toxicities, and X-ray attenuation properties. Iodine and barium are two contrast agents commonly used in fluoroscopic imaging, with K edges of 33 and 37 keV, respectively. The signal from iodine contrast is highly dependent on the X-ray spectrum used to image the contrast agent. The maximal contrast occurs when the polenergetic X-ray spectrum is optimized to be predominantly just above the K edge. However, the use of such low X-ray energies may lead to excessive patient dose, requiring careful selection of kV and appropriate filtration. As already introduced, the advent of high heat capacity X-ray tubes and high-power generators has made spectral shaping available. As many low energy X-rays that would contribute only to patient dose are removed, a lower kV can be used at the same patient dose rate, resulting in improved iodine contrast. The energy fluence of the X-ray beam is greatly
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reduced by the addition of Cu filtration, and the tube current must be increased to high levels (50–400 mA) to maintain acceptably short pulse widths. As patient thickness increases, the additional Cu filtration is gradually reduced to maintain short pulse widths and acceptable tube loading and to not increase the SNR by much. The noise level in fluoroscopic images is, in fact, already necessarily high, as a low IAKR is typically used to maintain the patient dose at an acceptable level.

Modern fluoroscopy systems have a number of software features that help reaching the achievable image quality. The use of edge enhancement software may improve visualization of vessels and other anatomic structures, and density equalization software has selectable parameters to reduce the contrast of bright areas, reduce flare, and boost the brightness of dark areas. Other software includes selectable parameters that modify the manner in which the contrast of imaged structures is displayed, a feature similar to a continuously adjustable characteristic curve for the display system.

21.3.3.2 Temporal Resolution

The excellent temporal resolution and its ability to provide real-time images are the advantages of fluoroscopy in comparison to radiography. However, both II and FPD systems exhibit image persistence or “ghosting”. Because the phosphorescent light of the scintillation surface undergoes a period of decay, light emissions from a previous image may persist as a “ghost” and degrade image quality. In FPD, this source of ghosting is limited by using a bright internal light source that flashes to reset the scintillation surface to background levels and an offset current.

Other than ghosting, a temporal blurring typically called image lag is also present. Lag implies that a fraction of the image data from one frame carries over into the next frame.

21.3.3.3 Noise

Fluoroscopy systems provide excellent temporal resolution, a feature that is the basis of their clinical utility. However, fluoroscopic images are also relatively noisy, and under certain circumstances it is appropriate and beneficial to (reduce) temporal resolution for lower quantum noise.

This can be accomplished by averaging a series of images. Different temporal averaging algorithms can be used, but a common approach to frame averaging is called recursive filtering. Recursive filtering is an image processing technique that combines portions of both the most recent fluoroscopic frame and several previous fluoroscopic frames to reduce noise in the resultant image. The recursive filter is thus a moving filter that incorporates information from several frames into the current fluoroscopic frame, reducing noise in the final image. Both quantum (X-ray) noise and additive noise from the video camera or image receptor are averaged.

The recursive filter works well if changes in the image from one frame to the next are small. In anatomical regions where motion is rapid, excessive recursive filtering can lead to unacceptable artificial lag. Artificial lag may also be noticed if instruments are moved rapidly or if the patient table is shifted. Most modern fluoroscopic systems use motion detection algorithms or other methods to prevent artificial lag. These algorithms monitor the change in image pixels from one frame to the next, and if the change exceeds a preselected threshold, the strength of the recursive filter is reduced until the image stabilizes. The strength of the filter also reduces fluoroscopic contrast. The compromise depends on the specific fluoroscopic application and the preferences of the user. Aggressive use of frame averaging can provide lower-dose imaging in many circumstances. Mobile fluoroscopy systems use X-ray tubes with limited output, and consequently temporal frame averaging is a common feature. In fixed C-arm systems, whereas some lag is usually beneficial for unrecorded fluoroscopic viewing, for recording dynamic events such as in DSA, lag is undesirable and recursive filtering modes are disabled.

In modern systems, recursive filters are no longer used and have been replaced by advanced processing that employs a different noise reduction function. This new algorithm promises to reduce the noise, without introducing any lag and, consequently, to improve the ability of accurately confirm the location and orientation of the catheter tip in the fluoroscopic images, without the need of using digital acquisition.

More advanced algorithms include real-time image processing. The image processing chain uses several features to improve image quality.

First of all, the real-time automatic pixel shift feature is used to reduce the anatomical structure noise, which is introduced in the subtracted image by patient motion or accidental table motion. By minimizing this undesired noise source, quantum noise will become the dominant noise source in DSA images.

Another feature is the temporal averaging of consecutive images to create a combined mask and a combined live image. Temporal averaging will reduce the amount of temporally uncorrelated noise, such as quantum noise. Contrast material detection functionality will demonstrate changes in the iodine bolus location and prevent this contrast material from being washed out by temporal averaging.

The third important feature of the algorithm is the spatial noise reduction. The first analysis phase aims to highlight the predominant signal structures on the image. These structures will be excluded from the low-pass spatial filter in the second phase. The combination of phases will only smooth the parts of the image that are considered featureless.

Benefits of these new algorithms include noise and artifact reduction, also on moving structures and object; image enhancement and edge sharpening; and automatic, real-time patient and accidental table motion correction on live images.

21.3.3.4 Spatial Resolution

The intrinsic limiting resolution of modern fluoroscopy detectors is quite high.

In II systems, the many signal conversions that occur degrade the sharpness of the fluorescent image. Electronic magnification (i.e., deminify the fluoroscopic image by selecting a smaller portion of the input phosphor to project on to the output phosphor) improves the spatial resolution, but at the same time decreases minification gain and decreases the sampling pitch of the input phosphor, thus increasing noise. To compensate, the technique factors are adjusted in order to maintain a constant perceived noise level in the displayed image. In an XRII, the IAKR usually increases as the ratio of the areas of the FOV as the image is magnified.
FPDs for fluoroscopy are designed to deliver similar or better spatial resolution as image-intensifier systems. In many systems, the full sampling resolution of the flat panel is maintained, independent of the displayed FOV. The video device and the effective sampling matrix size across the FOV, which determines the pixel dimension in the displayed image, mostly govern the limiting resolution of the entire system. With image zoom and panning, however, the full inherent spatial resolution of the acquired image can be obtained, regardless of the mode of operation. For systems with detector element binning, the intrinsic resolution is reduced by the binning factor.

Geometric magnification, achieved by moving the detector away from the patient, when used in conjunction with a small focal spot, can be useful in overcoming the spatial resolution limitations of a fluoroscopy system. However it necessarily increases the dose.

21.4 Dosimetric Consideration in Fluoroscopy

Millions of fluoroscopically guided interventional (FGI) procedures are performed annually, offering tremendous benefit over alternative invasive surgical procedures, faster recovery times, and improved patient outcomes and quality of life. As both technology and clinical practice continue to rapidly advance, more and more diagnostic and curative procedures are making use of fluoroscopic imaging. The growing use and increasing complexity of interventional procedures have been accompanied by a renewed focus on the management of ionizing radiation dose in order to ensure the highest safety for both patients and staff. Fluoroscopic procedures (particularly prolonged interventional procedures), in fact, may involve high patient radiation doses, which can give rise to both stochastic and deterministic effects (see Section IV, Chapter 66 of this book).

21.4.1 Radiation Safety Considerations for Patients

Stochastic risk is generally associated with an assessment of the increase in likelihood of the occurrence of cancers resulting from exposure to radiation. For any procedure involving the use of X-rays, the potential benefits of the procedure must be weighed against the stochastic risk. The exact magnitude of this risk remains under active debate for the relatively lower levels of radiation typically administered in medical imaging procedures. For interventional procedures, the risk/benefit balance is different from screening procedures, because the patient is known to have a serious medical condition that will very likely benefit from treatment.

The main radiation concern for patients in interventional procedures is usually deterministic risk, which is the tissue damage that will occur if radiation exposure exceeds certain thresholds. The tissue at greatest risk is usually the skin at the entrance location of the incident X-ray beam, where the intensity is greatest. Despite the fact that the number of these radiation injuries remains relatively small, they have a major impact on the patients who are affected. The skin response follows a specific pattern. The time period between the irradiation of the skin and the time that the effects appear as well as the threshold doses vary. The reactions can be prompt, early, midterm or long term.

- **Prompt reactions**: Prompt reactions occur less than 2 weeks after irradiation. A transient erythema appears from 2 to 15 Gy. Above 15 Gy, the erythema progresses into an acute ulceration that needs a surgical intervention.
- **Early reactions**: Early reactions occur from 2 to 8 weeks after irradiation. The effect is epilation from 2 to 5 Gy and epilation associated to erythema from 5 to 10 Gy. From 10 to 15 Gy, a dry or moist desquamation can also occur. Above 15 Gy, an epilation, an erythema and a moist desquamation appear.
- **Midterm reactions**: Midterm reactions occur from 6 to 52 weeks after irradiation. From 5 to 15 Gy, the epilation appeared in the first weeks (early reactions) can become partially permanent or even totally permanent. Above 15 Gy, there is a dermal atrophy, it corresponds to a secondary ulceration due to failure of moist desquamation to heal that needs a surgical intervention.
- **Long-term reactions**: Long-term reactions occur more than 40 weeks after irradiation. From 5 to 15 Gy, the erythema appeared in the first weeks (early reactions) can lead to dermal atrophy that needs a surgical intervention. Above 15 Gy, the dermal atrophy is more serious and needs surgery.

Fluoroscopically guided procedures can also result in high operator doses, and radiation safety is a critical component of a fluoroscopic imaging program. Patient and operator dose are strictly correlated, and optimizing patient dose means also reducing staff doses.

Radiation dose is optimized when imaging is performed with the least amount of radiation required to provide adequate image quality and imaging guidance. This process requires that all the factors affecting patient dose are taken into consideration and that actions are taken be taken before, during, and after a fluoroscopic procedures (Miller et al. 2010). A summary of actions to be taken is provided in Table 21.3.

21.4.1.1 Before the Procedure

Estimating the likelihood and severity of patient radiation effects requires consideration of:

- **Demographic factors**: It is well known those young patients are at higher risk of radiation-induced cancer compared to adults, since they are more susceptible

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<td><strong>Schematic Summary of Dose Management in Fluoroscopy</strong></td>
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to radiation dose and have a longer life expectancy. The risk is estimated to be three-times higher in young children compared to the risk of the adult population. For these reasons, in young patients the stochastic risk has to be considered of primary concern compared to the deterministic one.

Patient weight play also an important role in terms of radiation-induced risk due to the poor penetration, which is compensated by the equipment varying the exposure factors and, thus, increasing the exposure, and the closer proximity of the source to the patient. Figure 21.14 is a graphical representation of the variations of dose with patient size.

Figure 21.15 reports the measured values of Entrance Surface Air Kerma (ESAK) to a phantom of variable thickness for the three fluoroscopy modes (low, normal, and high) of protocol defined in a cardiac angiography equipment.

- **Medical history factors**: Certain clinical conditions, including genetic disorders, coexisting disease as diabetes...
mellitus and connective tissue disorder, use of medication and drugs and poor nutritional state, are suspected to pre-dispose patients to radiation-induced skin injuries.

Also, a recent high-dose procedure can result in the induction of effects at lower doses, depending on the radiation dose from previous procedures and the time interval between previous procedures and the planned procedure.

Pregnancy should prevent performance of fluoroscopy-guided procedures, except in case of critical conditions. In the latter case, all possible feasible expedients to minimize conceptus dose have to be applied.

- **Procedure factors:** Some types of interventional radiology procedures (i.e., neuro embolization, transjugal intrahepatic portosystemic shunt, complex cardiac interventions) are known to be “high dose” and, thus, to be associated with a higher risk for deterministic effects.

- **Equipment:** Procedures that may result in a clinically significant radiation dose should be performed by using fluoroscopic compliant with International Electrotechnical Commission Standard 60601-2-43. A medical physics expert should verify the appropriateness of measured exposure rates for typical clinical scenarios prior to the first clinical use of the equipment and repeat at least annually to ensure that patient radiation dose rates are consistent with those necessary to provide appropriate image quality.

- **Training:** Physicians performing fluoroscopically guided procedures should be trained in the safe use of fluoroscopic equipment in order to get familiar with dose-saving features of each type of equipment they use and the available tools to reduce patient dose and to understand the meaning of the dose indicators provided.

### 21.4.1.2 During the Procedure

Dose optimization is possible through appropriate use of the basic features of interventional fluoroscopic equipment and intelligent use of dose-reducing technology. Many technical parameters can be adjusted during the procedure to reduce radiation use or to improve image quality. These parameters include the choice of magnification, projections, and distances (Mahadevappa 2001).

As previously discussed, the use of magnification leads to an increase of radiation. The increment is $\propto 1/\text{FOV}^2$ for II systems and $\propto 1/\text{FOV}$ for FPD-based equipment. Therefore the lowest electronic magnification (largest FOV) required to perform the procedure has to be preferred.

Steep oblique or craniocaudal beam orientations increase the length of the radiation path through the body as compared with a posteroanterior (frontal) projection. As a result, these beam orientations require an increase in radiation output, sometimes by a factor of 10 or more, as compared with a posteroanterior projection (see Figure 21.16).

The proximity of the X-ray source to the entrance skin surface that is necessitated by these steep angles, according to the inverse square law, also results in an increase in skin dose (see Figure 21.17).

For the same reasons also the source-to-detector distance, for a fixed source-to-patient distance can play a role in reducing the dose and increasing the image quality, producing a sharper image. Positioning the image receptor as close to the body part as possible within the C-arm dramatically reduces patient exposure (see Figure 21.18).
Modifying the projection will not only reduce the patient dose, but also spread the radiation field on the patient’s skin reducing the peak skin dose, which is potentially responsible for any deterministic effect.

Peak skin dose reduction can be achieved also though the use of proper collimation. Even with dose spreading techniques, different irradiated fields can overlap on the skin surface. Tight collimation may prevent overlap (Figure 21.19). Modern systems support asymmetrical collimation.

Besides procedure optimization, operators are provided, in modern equipment, with various tools developed to help in reducing dose. The last image hold is an example. Last-image-hold is a feature that maintains the last fluoroscopic image on the viewing monitor while fluoroscopy is deactivated. It is achieved by continuously digitizing images in real time and temporarily storing them in a digital video frame memory. Last-image-hold allows the fluoroscopist to examine the image on the monitor for as long as necessary, using no additional radiation. Since in modern systems there is the possibility to store fluor images, the last-image-hold might lead to a reduced number of fluorographic (high-dose) images acquired. If a fluoroscopy loop provides adequate information for diagnosis or documentation, storing it instead of an fluorographic run can provide substantial dose savings.

Moreover virtual table movements, virtual collimation, and magnification can be operated on the last-image-hold without using additional fluoroscopy.

A more recent feature introduced is the spot fluoroscopy. This aims to reduce the exposed area during a projection. The operator can collimate asymmetrically around the detail of interest. Fluoroscopy will be performed only in collimated region, while the last-image-hold will be left in background to be used as reference.

Despite the optimization of the procedure and the use of dose reduction tools, high doses are sometimes unavoidable. For this reason, radiation dose should be monitored throughout the procedure, ensuring that the operator is aware of how much radiation dose is received by the patient. For this purpose, different dosimetric indicators have been used over the years to help operators in estimating patient dose, as fluoroscopy time, air kerma-area product, and reference air kerma.

Fluoroscopy time is the total amount of time during which fluoroscopy is utilized throughout the procedure. It still used as a surrogate for patient dose in fluoroscopy. It is, however, far from ideal, as it ignores many large contributions to patient dose, including digital acquisition imaging, which can be the largest contributor to patient dose during fluoroscopic procedures. Moreover it does not account for fluoroscopy rate, patient size, beam size, and beam position.

**Air Kerma-Area Product (KAP)** represents the total energy released to the patient. It can be directly measured using a KAP...
meter, or it can be calculated from known operating parameters. While KAP is an ideal quantity for assessing stochastic risk, it has limited application as an indicator of skin dose. In fact, it does not have the ability to distinguish between low doses over large fields and very high doses over small fields.

Reference air kerma (known also as cumulative air kerma) is the total dose accumulated during the procedure in a point explicitly defined by the manufacturer in respect to the radiation source, the isocenter of the equipment or the detector. For interventional equipment this point is called interventional reference point (IRP) and is located 15 cm below the isocenter towards the source along the source to detector axis. It should be representative of the position of patient skin during a standard cardiac procedure. However, the location of the IRP does not vary with changes in C-arm angle or focus to image distance (while the portion of skin irradiated does). Consequently, the dose is distributed over the skin throughout the procedure, while it is accumulated in the IRP. Furthermore, in a projection, the IRP may be outside the patient, may coincide with the skin surface, or may be inside the patient (Figure 21.20). As a consequence, the reference air-kerma might not be representative of the peak skin dose actually received by the patient. Lastly, the reference air-kerma is measured free in air and does not account for backscatter, table and pad attenuation, or tissue absorption.

Peak skin dose can be measured with some degree of accuracy using various dosimeters, as thermoluminescent dosimeters (TLDs) arrays and radiochromic film. However, measuring the peak skin dose for every patient is not feasible, and is extremely expensive and time consuming. Direct measurements have been instead used to determine specific trigger levels by studying the correlation between the peak skin dose measured and the dosimetric indicator (i.e., KAP and reference air kerma) in a sample of procedures. The purpose of these levels is to alert the operator about possible skin damage when the level is exceeded.

Some modern interventional equipment is now able to estimate skin dose estimation and to display it in real time on an intuitive and easy-to-interpret patient graphic (Figure 21.21).

21.4.1.3 After the Procedure

If a significant radiation dose has been administered, follow-up of the patient might be appropriate, as indicated by the ICRP publication no. 85. But even if the dose did not exceed the threshold for deterministic effects, recording patient dose is considered good practice. In the vast majority of countries, dose recording was a legal obligation by competent authorities before electronic tools like Hospital Information Systems (HIS), Radiology Information Systems (RIS), Picture Archiving and Communication Systems (PACS), or software tools for dose management were available. Hence, for many years paper forms tailored to the After dissemination of the DICOM (Digital Imaging and COmmunications in Medicine) standard, different DICOM objects have been widely used to store dosimetric information. The availability of standard repository objects and the increased awareness of radiation risk has led the development of patient-exposure-data monitoring solutions to support and facilitate the optimization of radiation protection of patients. The exposure-data-management systems further provide objective information to health care professionals and authorities who are responsible for ensuring justified and optimized use of radiation in medicine.

21.4.2 Radiation Safety Considerations for Operators

Unlike other modalities, staff dose in fluoroscopically guided procedures might represent an issue and raise concern. Operators and other personnel remaining in the procedure room during fluoroscopically guided procedures are exposed to scattered radiation and are at risk of developing both stochastic effects,
including cancer, and deterministic effects, namely cataracts. Occupational radiation protection requires appropriate education and training of the operators and the availability of appropriate protective tools and devices, as well as of adequate equipment.

Non-essential personnel should exit the room while fluoroscopy is activated and those persons remaining in the room should wear protective garments made of lead or an acceptable lead-free material. Mobile barriers are useful for reducing the radiation dose to persons who remain stationary during procedures, and suspended shields can be used to reduce the dose to the face, eyes, and neck of physicians while they are near the patient. It should be noted that the highest scatter radiation fields occur near the patient entrance field; therefore, standing closer to the image receptor is generally consistent with lower occupational dose levels.

REFERENCES

American Association of Physics in Medicine Report No. 125.


**FIGURE 21.21** Toshiba real time skin dose mapping. (ECR 2017 poster, courtesy of E. Vano.)