66
Radiation Protection Issues in X-ray Radiology, Fluoroscopy, and Computed Tomography

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66.1 Introduction: The Challenge of Radiation Protection in Radiology

Diagnostic X-ray imaging contributes to about 25% to 50% of the total (man-made plus natural) annual effective dose of the general population in Western Countries, and computed tomography (CT) is the largest single source of this medical exposure (UNSCEAR 2010; EC 2014) (see Section III, Chapter 38). In 2006, the average dose per inhabitant due to diagnostic radiology (excluding nuclear medicine procedures) in the US was 2.2 mSv, with 1.5 mSv of that total from CT examinations (NCRP 2010). In 2012–2013, the average dose per inhabitant in Germany, France, and Switzerland was equal to, respectively, 1.80 mSv (1.15 mSv due to CT), 1.47 mSv (1.14 mSv due to CT), and 1.42 mSv (1.00 due to CT) (BfS 2014; IRSN 2014a; Le Coultre et al. 2016), and, at the world level, the per caput effective dose increased by almost a factor of two (0.35 mSv to 0.62 mSv) from 1988 to 2008 (Shannoun 2015). Patients certainly benefit from these examinations, but in the last 20 years their impact on the collective dose has almost doubled and significant efforts remain to be made in order to control this trend and ensure that the benefit–risk ratio clearly lies on the benefit side. The collateral effect of X-ray imaging is patient exposure, which is associated with risks described in more detail in paragraph 666.1.1. In the low dose range of diagnostic radiology, the main risk is the stochastic one; especially cancer induction.

In that respect there are numerous papers published recently that include large epidemiological studies reinforcing the need for a strict application of the justification and optimization principles (Brenner 2002, 2007; Raelson et al. 2009; Pearce et al. 2012). If the absolute excess risks from single exposures remains low compared with background risks, at the moment, one cannot exclude an increase of cancer induction, due in particular to CT examinations in the pediatric population (Journy et al. 2017). Thus, the justification of an examination involving ionizing radiation should follow national or international guidelines. Concerning optimization of the radiological procedure, it is of primary importance to ensure that the level of image quality allows the clinical question to be answered while avoiding an unnecessary exposure of the patient. Fortunately, the level of patient exposure in radiology remains in general low. Nevertheless, overexposure incidents have been reported concerning, for example, CT brain perfusion examinations (Imanishi et al. 2005; FDA 2010). This confirms the importance of the use of quality assurance programs in radiology. Another area of concern when dealing with high levels of patient exposure is interventional radiology, where accidental overexposures have been reported (Coeytaux et al. 2015).

66.1.1 Radiation Protection in Radiology

Radiation protection ensures that ionizing radiations are used in a safe way. The International Commission on Radiological
Protection, ICRP, regularly publishes recommendations that are a key basis for international organizations and national authorities responsible for radiological protection (ICRP 2007a).

In its Publication 26, ICRP quantifies radiation risks and then proposes a system of dose limitation following its three principles of: justification, optimization of protection, and individual dose limitation (ICRP 1977). This recommendation was updated in 1991 (Publication 90) (ICRP 1977), classifying the radiation effects that result in tissue reactions as deterministic effects, and introducing the term stochastic effects for radiation-induced cancer and heritable disease. Sixteen years later another updated version was published (Publication 103), which took into account new scientific data available (ICRP 2007a).

National radiation protection legislations are generally based on ICRP Publication 103 and the Basic Safety Standards (BSS) published by the International Atomic Energy Agency, IAEA (IAEA 2014).

In spite of the fact that the effects of radiation on humans are quite complex, one generally distinguishes two main effects: the tissue reactions (or deterministic effects) and the stochastic effects. Tissue reactions are due in large part to the killing of cells following high doses. They are characterized by a threshold dose that varies from organ to organ and an increase in the severity of the reaction as the dose is increased. Their occurrence is not expected in standard radiological procedures. Nevertheless, they can occur during complex fluoroscopy procedures.

Stochastic effects (cancer or heritable effects) are due to mutations of somatic cells or reproductive (germ) cells. As opposed to tissue reactions, stochastic effects are considered as having no dose threshold (LNT: linear no-threshold model). The assumption is that the long-term, biological damage is directly proportional to the dose. For this effect, the probability of occurrence increases with the dose.

ICRP mentions that some radiation-associated health consequences, particularly some non-cancer effects (such as the development of cardiac pathologies), are not yet sufficiently well understood to assign them to either of the stochastic or tissue reaction categories.

Radiation protection in medicine is underpinned by the principles of justification and optimization. ICRP identifies justification as one of the cornerstones of radiation protection in medicine. Justification for medical applications acknowledges that exposures are used to help manage the patient. The justification process ensures that the benefits to the patient substantially outweigh any short- or long-term risks that the patient may incur. Justification is defined at three levels (ICRP 2007b; Malone et al. 2009, 2012):

- **Level 1**: The justification of the use of radiation in medicine is accepted as doing more good than harm. Its overall justification is taken for granted.
- **Level 2**: A specified procedure with a specified objective is defined and justified (e.g., chest CT for patients showing relevant symptoms). Level 2 should enable judging whether a radiological procedure will benefit those exposed.
- **Level 3**: Justification of a procedure for an individual patient. At this level, the application of the procedure to an individual patient must be justified (i.e., the particular application should be judged to do more good than harm to the individual patient).

The principle of optimization states that exposures should be kept at levels deemed sufficient for a correct diagnosis. The introduction of this principle relates to the adoption of the LNT model. This means that it is not enough to remain under the radiation dose limits. This principle is also known as the ALARA principle, which states that exposure to patients should be “As Low As Reasonably Achievable” for each procedure, given clinical need and patient factors.

With regard to medical exposure of patients, it is not appropriate to apply dose limits or dose constraints, because such limits would often do more harm than good. Some complex procedures might require a relatively high level of exposure and still be beneficial for the patient in comparison to alternative diagnostic or therapeutic options. An emphasis on the justification of the medical procedure and on its optimization should be applied. In order to limit the spread of the exposures applied for a given radiological procedure, the concept of Dose Reference Level (DRL) was introduced by the ICRP in its Publication 73 (ICRP 1996). ICRP Publication 105 confirms the fact that the use of DRL derived at the appropriate national, regional, or local level, is likely to be the most effective approach in an optimization process. The use of equipment features that facilitate patient dose management are also part of that Publication (ICRP 2007b).

Radiology has facilitated healthcare improvements, but recently the extent of its use has become a matter of concern for many reasons, including population dose, individual dose, budgetary and financial issues, and finally the appropriateness of the examinations or justification (Malone et al. 2012). To improve this situation, the concept of a clinical audit has been introduced (EC 2009; IAEA 2010).

The general justification of radiological examinations is outside the scope of the present chapter. We focus instead on the optimization aspects of radiology.

### 66.1.2 Compromise between Image Quality and Patient Exposure

A radiological image should be considered as a test to help patient management. The radiological image should convey the maximum amount of useful information, while minimizing patient exposure. If the quantification of patient dose is relatively easy, it is not the case for the image quality level that provides a successful and reliable diagnosis. Moreover, as shown in Figure 66.1, the efficacy of medical imaging can either be defined for single individuals, or up to a societal level, when obeying the following statement: “the demonstration of efficacy at each lower level in this hierarchy is logically necessary, but not sufficient, to assure efficacy at higher levels” (Fryback and Thornbury 1991, p. 88).

The first level focuses on the technical quality of the images; the second level addresses diagnostic accuracy using, for example, parameters such as the sensitivity and specificity of the technique used. Level 3 assesses whether the pieces of information obtained changed the referring physician’s diagnostic thinking; Level 3 is a prerequisite for Level 4 “efficacy,” which concerns the effect on the patient management plan. Level 5 assesses the...
effect of the information on patient outcomes, and, finally, Level 6 analyzes the societal costs and benefits of a diagnostic imaging technology. However, the efficacy of a radiological technique is in general assessed within the two first levels. Screening mammography might be an exception, where higher levels of efficacy are usually investigated. This type of examination is, however, focused on an asymptomatic population, as opposed to patients (see Section II, Chapters 19 and 25).

Within the optimization framework, one should separate two steps: the first is not only to ensure that the radiological unit is safe, but to ensure, also, that the primary photons transmitted by the patient (and, thus, carrying information) are optimally recorded by the image detector. The second step is to ensure that, for a given patient (considering his/her morphology and the diagnostic question), the protocol used complies with the ALARA principle. The first step is generally performed by the manufacturers, where image quality is evaluated as a function of the dose received by the image detector. To perform the measurements, one uses signal detection theory to obtain objective quantities. The way the spatial frequencies are transferred by the system can be characterized by the Modulation Transfer Function (MTF), the noise properties of the images produced can be characterized by the Noise Power Spectrum (NPS), the signal-to-noise ratio by the Noise Equivalent Quanta (NEQ), and the Detective Quantum Efficiency (DQE) can be used as a figure of merit (FOM) for the image detector (ICRU 1996). It is of note that DQE measurements alone are not sufficient to judge the performance of an image detector, since they only assess the way X-ray photons are converted into information, without considering the spatial correlation that might be introduced by the detector (due to the ratio of the MTF squared and NPS in the DQE formula). Thus, DQE results should be analyzed together with MTF assessments to get a full picture of the properties of an image detector. This kind of assessment corresponds to Level 1a, according to the various classes proposed in Table 66.1.

When dealing with flat panel detectors in radiography, standardized DQE or MTF measurements are made without an anti-scatter grid, but with beam qualities that are representative of X-ray spectra impinging the image detector in realistic clinical conditions. Thus, these measurements could be considered as not only reaching Level 1a, but also reaching Level 1b. When dealing with other situations, such as the use of scanning acquisitions with slits as scatter rejecting devices, or when willing to use the system without any rejecting scatter device, the standardized DQE or MTF measurements are no longer representative of the behavior of the system in a clinical situation. To reach Level 1b in such situations, the concept of effective DQE (eDQE) or generalized DQE (gDQE) have been introduced in which more realistic beam qualities are used together with the use of the scatter rejection device (Kyprianou et al. 2005; Samei et al. 2011; Salvagnini et al. 2013; Damet et al. 2014). This concept is certainly interesting, for example, to compare slit-scanning systems with flat panel systems, but there is no standardized method yet, and the problem of signal normalization needs to be clearly defined (for example, concerning the normalization with primary photons or primary and scatter photons impinging the image detector). This optimization step is fundamental and yet not entirely sufficient to ensure safe use of a system.

As a second step, one should balance the diagnostic information obtained with patient exposure; image quality then becomes task-oriented, and this task is generally performed by the end users (radiologists, radiographers, and medical physicists). Thus, to be most meaningful, image quality assessments should be related to actual clinical performance, which is difficult, expensive, and time-consuming. Furthermore, the results of such assessments can be strongly dependent on the patient sample and on the diagnostic tasks involved.

A practical, task-oriented image quality optimization scheme must necessarily involve limited criteria and use standardized methods of image quality assessment (level II in Table 66.1). Simple imaging tasks are thus, required for the benchmarking of image quality in radiology. The use of patient simulating phantoms can greatly simplify matters, but at the expense of case variability realism. Obviously, phantom details should mimic, to some degree, important disease-related structures in actual patients.

Both test object and anthropomorphic phantom images can be assessed using the receiver operating characteristic (ROC) paradigm or one of its derivatives (Localization ROC, Free-response

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**TABLE 66.1**

<table>
<thead>
<tr>
<th>Summary of Medical Image Quality Assessment Approaches</th>
</tr>
</thead>
<tbody>
<tr>
<td>- Level 1a: General image detector performances assessment (DQE and MTF assessment)</td>
</tr>
<tr>
<td>- Level 1b: General unit performances assessment (DQE or eDQE and MTF assessment)</td>
</tr>
<tr>
<td>- Level IIa: Task-based assessment: use of simple phantom and simple targets to be detected (use of ROC methods or mathematical model observers)</td>
</tr>
<tr>
<td>- Level IIb: Task-based assessment: use of anthropomorphic phantom with realistic targets to be detected (use of ROC methods or mathematical model observers)</td>
</tr>
<tr>
<td>- Level III: Clinically relevant assessment: use of actual images of patients with an assessment of the appearance of the normal anatomy (VGA methods) or with the assessment of lesion detectability (use ROC methods)</td>
</tr>
</tbody>
</table>

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ROC) (ICRU 2008). These methods give an accurate estimate of clinical image quality, but, although carefully controlled measurements, they are still subjective, since human observers are involved. These methods are time-consuming and require large samples and several observers to obtain precise results. In spite of these limitations these methods can be used either by radiologists (when dealing with clinical images) or naive observers when dealing with phantom images. As an alternative to ROC studies, one can use mathematical model observers that will try to predict human outcomes when dealing with simple tasks such as detecting a target in a homogeneous background (Myers and Barrett 1987; Beutel et al. 2000; Barrett and Myers 2004; He and Park 2013; Barrett et al. 2015).

The evaluation of the acceptability of clinical images cannot rely only on the assessment of phantom images. In radiology, human perception has to be taken into account and the relevant diagnostic content has to be balanced with patient exposure and not only with the dose received by the detector.

In 2000, Månsson proposed a four-grade scale to assess observer performances when reading radiological images, considering that an observer viewing and interpreting images should be able to (Månsson 2000; Seeram et al. 2014):

- Notice differences between varying states of disease
- Correctly describe diagnostically important structures and features
- Accurately categorize abnormalities
- Reliably distinguish relevant anatomical structures

It is interesting to note that these types of performance assessments fall into two categories: those based on lesion detection, where ROC type of studies can be performed, and those based on the visualization of anatomical structures, where visual grading analysis (VGA) can performed (Sund et al. 2004; Tingberg and Sjöström 2005; Seeram et al. 2014). This last approach makes it possible to reach Level III, according to Table 66.1.

Image quality assessed by these various approaches is dependent on detector dose. However, the link between detector dose and patient dose is not straightforward. Usually, manufacturers will optimize their system with respect to detector dose. The end user of the unit should, nevertheless, optimize the use of the unit on patients with respect to patient exposure. Because of this, medical physicists should be involved not only to ensure the compliance of a radiological system with legal requirements, but also to ensure that patient optimization is achieved.

### 66.2 Radiation Protection in Radiography

#### 66.2.1 Technical Requirements

Radiography can be thought of as photography performed with X-rays. The exposed anatomy is projected on a plane; images were recorded in the past on film, which is now being replaced by digital media. The complete radiological chain is made of the following components:

- A high voltage generator
- An X-ray tube
- An image display device
- An anti-scatter grid
- An automatic exposure control (AEC) device
- An image detector

To ensure the safety of a radiography unit, standards have been published concerning the characterization of the various components of the radiography unit, and regulatory authorities have set limits following national requirements such as the ones set by the Food and Drug Administration (FDA 1997) or international standard (IEC 2009a, 2010a), or recommendations such as those proposed by the European Commission (EC 2012). This section will make a brief review of the main standards or recommendations available.

The high voltage generator of an X-ray unit also provides the circuitry and electronic control for the filament X-ray tube current, the rotation of the anode, as well as the exposure time. One of the most important parameters in radiography is the mean X-ray beam energy that is responsible for the contrast of the radiant image (modulation of the primary X-ray beam) obtained from the differential attenuation of the tissues. The variation in current and high voltage across an X-ray tube, which occurs over the voltage waveform cycle, produces changes in the photon flux and spectrum shape (Birch et al. 1976; Seibert 1997). To control image contrast and patient exposure, it is necessary to verify whether the performance of the generator complies with its expected characteristics. Modern kV-meters make it possible to assess the high voltage average and peak values, and also often enable the display of the whole waveform to estimate the ripple factor.

Concerning the X-ray tubes, there are several standards enabling their full characterization. Image quality not only depends on contrast, but also on spatial resolution, something which mainly depends on the size of the focal spot. This parameter can be evaluated using an IEC (International Electrotechnical Commission) standard (IEC 1993). A minimum beam filtration, generally expressed in equivalent millimeters of Aluminum, is required to avoid overexposing the patient’s skin (FDA 1997). The usual requirement concerning that parameter is at least an X-ray beam filtration of 2.5 mm eq. Al. To verify if the X-ray tube complies with this requirement, one generally measures the Half-Value Layer (HVL) of the beam and from that measurement estimates the total filtering of the beam (DIN 1990; IPEM 1996). The adequacy between the field of view shown by the collimator light field and the actual X-ray field is also defined (EC 1997). Finally, tube leakage should comply with the limit generally adopted of 0.88 mGy/h at 1 m (FDA 1997; IEC 2008a).

The table that supports the patient has to be stable and made of a low attenuation material free of structures that would introduce a structured noise in the radiograph. As mentioned by Aichinger et al. (2012), it is quite surprising that, in spite of the importance of that parameter, minimum transmission factors are very difficult to find, and one can wonder if they actually exist.

Anti-scatter grids improve image contrast by absorbing scattered radiation produced by the patient, but increase patient dose when willing to keep detector dose constant. Their performances are standardized, but their actual scatter rejection properties are not.
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remain difficult (IEC 2003a). Moreover, the characteristics one might get from these standardized methods are, in general, far from what one actually gets when dealing with patients. In such situations it seems that the use of a standardized effective DQE could be very useful for benchmarking radiography units.

When dealing with an image detector with limited latitude of exposure, such as the screen-film system, reproducibility of the response of the automatic exposure control (AEC) is a very important aspect. Moreover, its adjustment has to take into account the energy dependence of the screen-film system and the non-reciprocity properties of film. With digital detectors, in spite of their large latitude of exposure, reproducibility of the AEC system still matters to ensure a constant signal-to-noise ratio. Usually one checks if the reproducibility of the response is within a given tolerance. The image detector is obviously a very important part of the whole imaging chain, and its characterization has always been extensively described. When dealing with screen-film systems the exposure required to get an image is defined by the sensitivity or “speed” (ISO 2001). When dealing with digital detectors, the user can select the level of sensitivity he or she wants to achieve. To indicate the exposure level of the detector, several manufacturers have introduced exposure indicators or exposure indices (IEC 2008b). Unfortunately, there is not yet a unified exposure index, even for flat panel type image detectors. Nevertheless, the image quality performances of digital detectors can be objectively assessed using the normalized DQE parameter (IEC 2003b). Finally, the device used to read the radiological image should comply with DICOM 3.14 standard (DICOM 2011). Quality assurance programs should then be applied to ensure constancy of the performances over time (AAPM 2002). Several test objects, such as the ones presented in Figures 66.2 and 66.3, are available to evaluate the overall quality of the radiographs produced by the unit.

66.2.2 Optimal Way to Use the Unit

Compliance with the requirements mentioned in the previous section ensures the safety of radiography units. However, major efforts have to be made to ensure that the units are then used in a safe way. DQE measurements, image quality assessment–Level Ia, according to Table 66.1, are performed in beam qualities representative of standard radiology, and are excellent figures of merit to judge the performances of an image detector. However, they only deal with one element of the imaging chain. The DQE does not include the scatter rejecting device or the patient support. In order to get closer to a more clinically relevant geometry, the concept of eDQE or gDQE was introduced (Kyprianou et al. 2005; Samei et al. 2009, 2011; Bertolini et al. 2012; Clavel

FIGURE 66.2 Example of test object available for quality control of digital X-ray units. (NORMI 13 X-ray Test Object from PTW German.)
et al. 2016). This quantity, image quality assessment—Level Ib, according to Table 66.1, is unfortunately still not standardized. Finally, if this parameter makes it possible to create a benchmark of radiography units, it deals with a dose measurement performed at the entrance of the image detector, or detector + anti-scatter grid for eDQE/gDQE, and not patient dose.

At this stage, it is important to note that DQE or eDQE/gDQE should be performed on minimally processed images; this leads to image characteristics that might be quite different from the actual clinical ones.

In addition, and as mentioned at the beginning of this chapter, a better way to define image quality would be to first define a task. In radiology, there are essentially two main tasks: the detection of very small structures having a high contrast level with the background of the image, and the detection of relatively large structures (5–8 mm) of very low contrast. The spatial resolution characteristics of the image will be very important when dealing with small structures, whereas image noise (and, thus, entrance air kerma to the detector) will condition the detection of low contrast structures.

Bar patterns can be used to assess the characteristics of spatial resolution, but one has to be aware of the data sampling of the digital system. In this sense, bar patterns should be placed at 45° with respect to the pixels matrix. Pixel size, detector air kerma, and DQE are key parameters when dealing with image noise. In low dose mode, one should check if the manufacturer has not introduced some pixel binning, something which drastically reduces image noise, but with an associated significant loss of spatial resolution.

Concerning the evaluation of the low contrast detection, it can be performed with test objects such as the Contrast Detail and Conventional Radiography (CDRAD) phantom that can be scored subjectively by human observers or using a computer program that computes a single image quality figure of merit (IQF$_\text{th}$) (Artinis 2000; Pascoal et al. 2005). One can add slabs of Poly(methyl methacrylate) (PMMA) to simulate the thickness of a given anatomical region and optimize the image quality outcome with respect to not only the air kerma at the entrance of the detector, but also the air kerma at the entrance of the beam, thus including patient exposure in the process. This type of image quality corresponds to Level IIa according to Table 66.1. Keep in mind that, for a given level of entrance air kerma to the detector, beam energy variations (i.e., kV) will have no major impact on results, since this test object does not include high Z materials.

Some authors have proposed other types of phantoms, such as the one presented in Figure 66.4, reaching the image quality assessment of Level IIb according to Table 66.1 (Mah et al. 2001). Approaches implying the use of mathematical model observers have also been proposed (Ullman 2008).

In 1996, the European Commission published a set of guidelines concerning the most frequently performed radiography using the image quality assessment of Level III according to Table 66.1 (EC 1996a,b). In these guidelines, not only were technical factors proposed, but image quality criteria based on the appearance of normal anatomy were also defined. Image quality optimization could then be performed using techniques such as the Visual Grading Analysis (VGA) to assess clinical images.
Radiation Protection Issues in X-ray Radiology, Fluoroscopy, and Computed Tomography (Tingberg and Sjöström 2005). This initiative made for a drastic reduction of the variability in image quality and patient exposure. Unfortunately, the proposed technical parameters are for screen-film systems, and no update of such a document is yet foreseen. The International Commission on Radiation Units and Measurements, ICRU, has, however, published a report specifically on chest radiography, since it is the most frequently taken radiograph in the world (ICRU 2003).

Radiography optimization focusing on image quality is the most appropriate way to go, but it is also the most difficult one. To limit the spread of the practice, the concept of DRL was first mentioned by the ICRP in 1990, and subsequently recommended in greater detail in 1996 (ICRP 1991, 1996). These levels are not the suggested or ideal dose for a particular procedure, or an absolute upper limit for dose. Rather, they represent the dose level at which an investigation of the appropriateness of the dose should be initiated. DRL for radiography of adults can be found in various publications (EC 1999; Freitas and Yoshimura 2009). Concerning the pediatric population, particular efforts have to be made in terms of optimization, as this population has a higher risk than adults of developing cancer (BEIR 2006). As was done for adults, in 1996 the European Commission published recommendations concerning clinical image quality requirements, with representative values of entrance surface dose for various children age categories (EC 1996b). In early 2011, the European Commission launched a new project, a follow-up called “Study on European Population Doses From Medical Exposure” (EC 2008), or Dose Datamed 2 (DDM2) to update the available dose values from medical exposure procedures, in X-ray diagnostics, interventional radiology, and nuclear medicine (NM), in the European Union Member States. One of the results of this survey showed that, for pediatric radiology, national DRLs could vary by a factor 7 (EC 2014). Because of this the European Commission recognized that some efforts were needed to consolidate the available data, and launched, in 2013, the PiDRL project, also supported by the European Society of Radiology (ESR) in its EuRosafe initiative (ESR 2014). In the United States, a similar initiative was launched by “The Alliance for Radiation in pediatric imaging—Image Gently” (IG 2015). A similar approach was taken for adults under the name of “Image Wisely” (IW 2015). As a final remark that applies to the whole field of X-ray imaging, DRLs are certainly a way to reduce the spread of the practice, but they should be given for specific indications instead of anatomical parts. There is a risk that the present general trend of lowering patient exposure might result in producing suboptimal image quality, thus impairing patient care. Clinical image quality in radiology always needs to be at the front.

66.3 Radiation Protection in Fluoroscopy
Fluoroscopy, also called radioscopy, is a real-time X-ray imaging modality (see Section II, Chapter 21). Based on a similar imaging chain to radiography, it facilitates the visualization of various dynamic phenomena in the human body. This makes it possible not only to diagnose specific diseases, but also to treat them immediately or shortly thereafter, as is depicted in Figure 66.5. As such, it is used in a large variety of medical specialties, such as GI tract imaging (e.g., barium meals, barium enemas), orthopedic surgery (e.g., fracture reduction), vascular interventions (e.g., coronary blood vessel repair, transjugular intrahepatic...
portosystemic shunt (TIPS) placement, neuroangiography, or percutaneous ablation procedures (e.g., arrhythmia healing, renal sympathetic denervation for the treatment of resistant hypertension). As a consequence, exposure times for patients may reach between a few seconds to a few hours of beam-on time, depending on numerous factors such as patient health condition, vascular abnormalities, unexpected hemorrhage, or operator experience. This leads to potentially high patient doses, especially skin doses.

The use of fluoroscopy in medicine is ever increasing, due to many upsides of the technology. For example, the patient hospitalization time for a percutaneous coronary intervention (PCI) is on the order of a few hours compared to several days for open-heart surgery. Furthermore, the minimally invasive nature of percutaneous interventions drastically reduces septic risks for the patients, thus reducing potential costs. Finally, the outcome of these less invasive procedures often shows higher success rates. For example, the fraction of ST (segment of the electrocardiogram, ECG) Elevation Myocardial Infarction (STEMI) patients undergoing systemic drug-induced thrombolysis went from 51% (male) and 39% (female) in 1997 to less than 1% for both genders in 2007. On the contrary, the fraction of PCI treatments rose from under 10% for both genders to 80% (male) and 70% (female) in the same period of time (Radovanovic et al. 2012).

These elements imply that the use of fluoroscopically-guided interventions is intrinsically more irradiating for patients. Miller et al. (2009) showed this in their proposal for reference levels (RL) for interventional radiology. As can be seen in Table 66.2, some interventions involve doses that are de facto high, such as brain tumor embolization, with a 75th percentile of the air kerma-area product (P_{K,A}) distribution of 403.21 Gy cm² and cumulative air kerma (K_{a,r}) at 4.169 Gy.

For many of these interventions, a medical staff member is needed in the close vicinity of the patient during image acquisition. The immediate consequence of this is a potentially high occupational exposure, which is, in turn, highly dependent on the position of the staff in the operating room (ICRP 2013).

### 66.3.1 Technical Requirements

A crucial element in the imaging chain is the AEC system, also known as “Automatic Dose Rate and Image Quality Control” (ADRIQ) or “Automatic Brightness Control” (ABC). Indeed, the selected power curves for a given acquisition protocol will automatically determine the X-ray tube voltage and current, as well as the beam filtering, on the basis of the detector dose rate. As a consequence, this system should be carefully calibrated.

<table>
<thead>
<tr>
<th>Procedure</th>
<th>No. of Cases of Procedures</th>
<th>Reference Dose (Gy)</th>
<th>KAP (Gy cm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>75th Percentile</td>
<td>95% CI</td>
</tr>
<tr>
<td>Transjugular intrahepatic portosystemic shunt creation</td>
<td>134</td>
<td>2.162</td>
<td>1.689–2.502</td>
</tr>
<tr>
<td>Biliary drainage</td>
<td>123</td>
<td>0.965</td>
<td>0.767–1.195</td>
</tr>
<tr>
<td><strong>Nephrostomy</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>For obstruction</td>
<td>76</td>
<td>0.272</td>
<td>0.226–0.342</td>
</tr>
<tr>
<td>For stone access</td>
<td>62</td>
<td>0.406</td>
<td>0.283–0.549</td>
</tr>
<tr>
<td>Pulmonary angiography</td>
<td>104</td>
<td>0.293</td>
<td>0.255–0.341</td>
</tr>
<tr>
<td>Inferior vena cava filter placement</td>
<td>274</td>
<td>0.146</td>
<td>0.128–0.169</td>
</tr>
<tr>
<td><strong>Renal or Visceral Angioplasty</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Without stent</td>
<td>53</td>
<td>1.082</td>
<td>0.880–1.487</td>
</tr>
<tr>
<td>With stent</td>
<td>103</td>
<td>1.552</td>
<td>1.396–1.693</td>
</tr>
<tr>
<td><strong>Iliac Angioplasty</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Without stent</td>
<td>24</td>
<td>1.015</td>
<td>0.805–1.187</td>
</tr>
<tr>
<td>With stent</td>
<td>93</td>
<td>1.316</td>
<td>1.008–1.629</td>
</tr>
<tr>
<td><strong>Bronchial artery embolization</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Without stent</td>
<td>27</td>
<td>1.312</td>
<td>1.041–1.699</td>
</tr>
<tr>
<td>With stent</td>
<td>150</td>
<td>2.472</td>
<td>2.050–3.134</td>
</tr>
<tr>
<td><strong>Hepatic chemoembolization</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Without stent</td>
<td>125</td>
<td>1.448</td>
<td>1.348–1.548</td>
</tr>
<tr>
<td>With stent</td>
<td>103</td>
<td>1.699</td>
<td>1.390–2.143</td>
</tr>
<tr>
<td><strong>Uterine fibroid embolization</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Without stent</td>
<td>88</td>
<td>1.699</td>
<td>1.390–2.143</td>
</tr>
<tr>
<td><strong>Other tumor embolization</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Without stent</td>
<td>94</td>
<td>2.056</td>
<td>1.797–2.599</td>
</tr>
<tr>
<td><strong>Gastrointestinal hemorrhage localization and treatment</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Without stent</td>
<td>134</td>
<td>5.34</td>
<td>4.615–5.937</td>
</tr>
<tr>
<td>With aneurysm</td>
<td>148</td>
<td>4.441</td>
<td>3.960–5.239</td>
</tr>
<tr>
<td>With atheroma</td>
<td>51</td>
<td>4.169</td>
<td>3.479–5.070</td>
</tr>
<tr>
<td>With endoleak</td>
<td>98</td>
<td>1.848</td>
<td>1.479–2.243</td>
</tr>
<tr>
<td>Pelvic artery embolization for trauma or tumor</td>
<td>35</td>
<td>1.874</td>
<td>1.749–2.167</td>
</tr>
<tr>
<td>Embolization in the spine for AVM or tumor</td>
<td>21</td>
<td>5.072</td>
<td>3.453–7.456</td>
</tr>
</tbody>
</table>

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by the manufacturer and assessed by the medical physicist. The American Association of Physicists in Medicine (AAPM) issued a report on the matter in 2012 (AAPM 2012).

Due to the high exposures related with fluoroscopic guidance systems, a reliable dose management program, along with precise dose indicators, should be available on all fluoroscopy units (ICRU 2005). The four main dose indicators that should be implemented are:

- Fluoroscopy time (T)
- Number of radiographic frames per intervention (N)
- Cumulative air kerma-area product ($P_{KA}$)
- Cumulative incident air kerma ($K_{air}$)

The fluoroscopy time (T) is defined as the time during which the pedal is being pressed by the operator. Since the X-ray beam is usually operated in pulsed mode, this value is far from being indicative of the actual patient exposure time. Let us take a simple numerical example: the typical pulse width is of the order of 10 ms. At 15 fps, the actual beam-on time will only be 15% of the fluoroscopy time. Furthermore, some manufacturers measure only fluoroscopy time, thus not incrementing the chronometer when proceeding with diagnostic quality imaging, such as digital subtraction angiography (DSA). As a consequence, T is indicative of the complexity of the intervention, but is not an ideal dose indicator to estimate patient dose.

The number of frames per intervention (N) is, like the fluoroscopy time, an indicator of the complexity of the intervention—or even an excessive frame acquisition rate, sometimes not suited for a given diagnostic purpose—but does not provide an accurate estimation of patient dose (Jones and Pasciak 2011). Indeed, although much information about the acquisition parameters (tube voltage and current, pulse width, focus-to-detector distance, table height, etc.) is stored in the respective Digital Imaging and Communications in Medicine (DICOM) headers, many images are usually not recorded, especially the low-quality fluoroscopy frames used solely for positioning or navigation purposes. Only diagnostic quality images are usually recorded. A solution for this problem could come from a new DICOM object, the Radiation Dose Structured Report (RDSR), recording many operational parameters for all exposure events.

The cumulative air kerma-area product ($P_{KA}$) is an interesting dose indicator of patient exposure. It is defined as the integral of the air kerma cumulated at a given location by the area exposed by the primary beam at that same location. Incidentally, the $P_{KA}$ does not depend on where it is measured. Indeed, for a point-like source (as can be hypothesized of the focal spot), the air kerma rate decreases with the inverse square of the distance to the focal spot, whereas the exposed area of a cone-shaped beam increases with the square of the distance to the focal spot. As such, the integral (or product) of both is an invariant with respect to the location in the primary beam. It can, thus, be easily converted to patient effective dose using conversion factors determined by extensive Monte Carlo simulations (Struelens et al. 2009).

For a given medical purpose, the X-ray beam will usually be focused on a given anatomical region (head, chest, pelvis, etc.) using beam angles as defined by different medical requirements. The organs in the primary beam, along with those exposed by internally scattered X-rays, are, thus, more or less the same, so average conversion factors can be established for a quick estimation of patient effective dose. For example, suppose a pelvic artery embolization performed on a male patient yields a cumulative $P_{KA}$ of 250 Gy cm². Admitting that the beam was oriented in the postero-anterior (PA) direction, if the HVL of the beam is measured to be 5.5 mmAl (DIN 1990; IPEM 1995), the corresponding conversion factor (Struelens et al. 2009) is 0.109 mSv/Gy cm². Thus, the estimated effective dose will be 27 mSv. If one does not have the time to estimate the HVL, a conservative estimate would be to take the highest HVL, thus overestimating the effective dose.

Finally, the cumulative incident air kerma, $K_{air}$, unlike the $P_{KA}$, has to be defined at a given relevant location. The IEC (IEC 2010b) defines the interventional reference point (IRP) as virtually lying at 15 cm from the device’s isocenter towards the X-ray tube along the central beam axis. The IRP should be representative of the entrance point of the X-ray beam in the patient. As such, the $K_{air}$ will be used as an estimator and/or predictor of possible patient skin injuries. For this purpose, one should use a substantial radiation dose level (SRDL), defined in Balter et al. (2011) as “a facility-selected value used to trigger patient follow-up for possible deterministic injury. (p. 1613)” Usually, the SRDL for cumulative air kerma is set at $K_{air} = 5$ Gy, yielding a peak skin dose (PSD, maximum skin dose of a patient after a procedure) of roughly 3 Gy.

The FDA reference point differs from the IEC reference point, in that it is defined at 30 cm from the entrance of the imaging device, thus moving along the main primary beam axis. As such, the $K_{air}$ recorded at either reference point, even with a fixed geometry during the intervention, may over- or underestimate the actual PSD, depending on patient body habitus (Balter 2006). As a consequence, the $P_{KA}$ indicator of the fluoroscopy device should be within a tolerance margin of ±35% according to the FDA, and even ±50% according to the IEC (Balter 2006).

66.3.2 Operational Requirements

The most fundamental starting point in patient and staff dose optimization is a good education and knowledge of unit behavior when it is clinically used. The acquisition protocol has to correspond to the desired clinical purpose and anatomical region. Indeed, the AEC system rests on a set of so-called power curves, as shown in Figure 66.6 (adapted from AAPM 2012). In order to adapt the air kerma rate to the detector (or Entrance Exposure Rate to the Detector [EERD]), the AEC will increase both the tube voltage and current until it reaches a detector entrance dose determined by the sensitivity pre-setting. High detector sensitivity will induce a low dose rate, whereas higher quality images will lower the sensitivity and increase the detector dose rate.

The respective power curves have been calibrated to yield an acceptable image quality for a given anatomical region and a different quality level. As such, the use of a protocol for a given anatomical region instead of another may lead to inherent image (and patient) under- or overexposure, potentially impinging on the diagnostic task.

When the right protocol has been chosen, the operator has to know which parameters play a major role on the image quality and also the patient dose, such as frame rate, fluoroscopy or fluorography, beam incidence, and patient position in the
primary beam. There are no straightforward rules as to how the dose rate adapts to the different input parameters (patient body habitus, beam angle, field size, magnification setting, detector sensitivity, etc.) (Vaño et al. 2009). In order to get a better understanding of the influence of those parameters, a good—although time-consuming—way to sort out their respective influences is to measure the device’s dose rate using controlled conditions, such as a dosimetry system (semi-conducting or ionization chamber) and a phantom (anthropomorphic or PMMA slabs) that simulates the patient. The dose rate that will be measured at the entrance point of the beam in the slabs, containing a given fraction (30% to 40%) of backscattered radiation, is known as the Skin Entrance Exposure Rate (SEER). The measurements can be performed by a device user, a radiation protection expert, or a medical physics expert. An example of the results can be seen in Figure 66.7.

This kind of database can serve as a basis to optimize one’s clinical practice, by making it possible to alter a directly accessible operational parameter that may have an influence on patient exposure on a given unit, but may not have any influence for another manufacturer.

### 66.3.3 Optimal Way to Use the Unit

Due to its use for interventional procedures, radiation protection for fluoroscopy is of concern for two different populations. First, the patient is primarily affected by the detrimental effects of ionizing radiation, and, because these interventions can last between a few minutes to several hours, the span of risks is quite large. Where most diagnostic procedures deal with the minimization of stochastic risks, because of its high skin doses, fluoroscopy is also largely concerned with tissue—and especially subcutaneous—lesions. The second at-risk population includes device operators and, on a larger scale, the staff present in the operational theater around the patient. Indeed, the main source of exposure in interventional procedures is the radiation scattered by the patient, mainly because of Compton scattering in the patient’s soft tissue.

#### 66.3.3.1 Patient

As mentioned previously, the patient is exposed to stochastic effects (to be minimized) and tissue effects (to be prevented, or at least taken care of when they occur) (Balter et al. 2010). In this sense, the aim is to keep these risks to a minimum by, respectively, keeping the $P_{K_a}$ and the $K_{a,r}$ at a minimal level.

It is important to note that fluoroscopically-guided procedures can present risks for skin injuries (tissue effects, proportional to $K_{a,r}$), as well as an enhanced risk of long-term cancer development (stochastic risks, proportional to $P_{K_a}$). To facilitate the approach on radiation protection for the patient for fluoroscopic examinations, one can separate both paradigms. Stochastic risks should be minimized for procedures intrinsically yielding low doses: Digestive series, arthroscopy, orthopedic procedures, and so on, as well as pediatric examinations. Indeed, for the latter, the body habitus of infants and children is thin enough that dose rates to the skin remain low. Furthermore, special protocols are usually dedicated to pediatric procedures. The accent will, thus, be placed on minimizing the long-term complications due to high effective doses. It is to be noted that many mobile fluoroscopy units are often used outside a hospital’s imaging department, and might be used by insufficiently trained staff (ICRP 2010).

Interventional procedures, such as coronary angiography or cerebral aneurism treatment, especially for adults, still present stochastic risks proportional to the $P_{K_a}$, but the main risk to be addressed is skin injuries. For these procedures, which yield intrinsically high effective doses, great care should be taken to reduce the PSD at an acceptable level by accepting lower image quality—thus lower dose rates—or changing the beam angle regularly so as to “spread out” the entrance $K_{a,r}$ across a larger area (ICRP 2013). The first step in optimizing patient dose is to choose the right protocol, adapted to the region to explore and to the patient size. Ideally, the default parameters should be set to minimal quality requirements, which can be manually set to a higher level, instead of immediately starting with the best image quality, perceptually harder to tune down. As mentioned by Sund
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et al. (2004), “the visibility of normal anatomy is strongly correlated to the detectability of pathological structures. (p. 49)” A common pitfall is reducing the dose to such an extent that the diagnostic task becomes difficult, implying perverse side effects such as increased volume of injected contrast medium, whose renal toxicity is proven and immediate. To prevent this, image quality should be assessed with respect to a given detection task. This can either be performed by assessing diverse phantom images by several human observers, or the implementation of a model observer (Favazza et al. 2015), in order to ensure minimal dose requirements for a given task. The key to optimizing patient dose—aiming to avoid excessive patient dose and ensure maximal diagnostic ability—is, thus, to properly distinguish normal anatomy. To do this, then, the main elements playing a role in patient dose optimization are the following:

- **Focus-to-skin distance**: Ideally, the patient should be kept as close as clinically possible to the imaging device and as far as possible from the X-ray tube. The beam being divergent, patient skin dose will increase with decreasing distance to the tube. Incidentally, this also has an effect on image quality, since the penumbra from the non-punctual focal spot is lessened with increased focus-to-patient distance (by reducing the geometric magnification effect of a conical beam).

- **Tube angle**: For a given anatomical region, the entrance skin dose rate will increase exponentially with tissue thickness. This is mainly due to the AEC system, which ensures a constant dose rate at the detector for a given sensitivity. Thus, the beam angles that minimize cross-patient thickness will lower the skin exposure. A quick method to estimate the increase in skin dose rate is to keep in mind that the HVL of soft tissue at 70 kV and for a 3 mm Al filter is 3 cm. For example, let us suppose a patient is 15 cm thick in the postero-anterior (PA) direction and 30 cm in the lateral direction. This represents an increase of 15 cm or 5 HVL of soft tissue. The increase in dose rate will be approximately \(2^5 = 32\). This also means that a substantial gain on skin dose can be achieved by removing any unnecessary body parts from the primary field (e.g., by asking the patient to put his/her arms up during a chest intervention using lateral or oblique projections).

- **Primary beam collimation**: On a fluoroscopy unit, there is at least one primary collimator, usually made of lead. Depending on the device, the collimator can be circular (iris), parallel using two blades, or rectangular using four blades. This device makes it possible to close the primary beam on a given part of the anatomy. This has an effect on patient dose (by reducing the exposed area) and image quality (by reducing the scatter fraction). For example, a 20’ × 20’ flat panel detector is too large for a cervical spine examination. The use of two collimation blades can blind out the anatomy that does not have to be irradiated. Furthermore, the dynamic difference in the image is lessened, since the highly absorbing vertebral show a large contrast with respect to the surrounding tissue or even free air.

The human eye being only sensitive to around 20 levels of gray, too high a contrast is not needed, and could be potentially harmful, for the correct diagnosis or treatment. A further set of collimators, usually present on the larger devices, is the so-called “soft filter.” This is made of aluminum plates that can be moved in different parts of the anatomy, in order to reduce highly different contrasted tissues. For example, typical cardiac exploration protocols automatically slide part of such a collimator in the lung region, so as to reduce inherent contrast between air and soft/bone tissue, for the same reasons as mentioned previously.

- **Electronic magnification**: The use of electronic magnification usually increases patient dose. In order to zoom in on a specific part of the anatomy, the device closes part of the primary collimation system. On a luminance amplifier, this implies a smaller irradiated surface, and fewer electrons produced in the photocathode. The terminal charge-coupled device (CCD) being the input for the AEC system, the latter will increase patient dose, roughly by the ratio of both exposed surfaces. For flat panel detectors, this also implies a smaller irradiated surface, but the mechanisms behind the increase in dose are more subtle. Indeed, the physical pixels are usually more numerous than the displayed ones, thus implying pixel binning. In order to keep the same image matrix on the display screen, less physical pixels are binned together to form a single image pixel. Thus, to keep the noise level constant, the dose has to be increased.

After the intervention, the patient dose—in terms of \(P_{\text{KA}}\) and \(K_{\text{A}}\)—should be assessed and used for patient follow-up. Indeed, tissue effects arise with typical latency periods of several hours to several days, depending on the actual dose. This means that, when exceeding a given SRDL (Balter et al. 2011), the patient should be advised to regularly check the area where the beam entered his/her body, and where the effects are most likely to happen. If anything unusual occurs (skin reddening, hair loss, etc.), the patient should contact the physician who performed the examination. After a more precise dose estimate—on the basis of the dose indicators—the patient should be taken care of, ideally by a radiation oncologist, who is more likely to be familiar with radiation skin injuries.

### 66.3.3.2 Staff

The primary source of operator and medical staff exposure is the patient. Indeed, part of the primary beam undergoes Compton scattering, mainly in the soft tissue. This creates stray photons of typically lower medium energy with respect to the primary beam, which will irradiate the surroundings (McCaffrey et al. 2012). The intensity of scattered radiation fields can reach several tens of mGy/h, thus necessitating protective clothing (lead aprons and thyroid collars), since most of the sensitive organs are in the trunk area. This habit is now firmly implanted in interventional radiology; however, it calls for extra care to be taken in order to ensure the physical integrity of leaded clothing. Indeed, these items are usually made of metallic dispersions in polymer
materials (McCaffrey et al. 2012). These materials, when regularly folded, can develop cracks in their matrix, leading to unaware staff exposure. A good and practical storage solution for lead aprons is of prime importance.

Furthermore, in recent years, there has been an increasing concern about operator brain and eye lens exposure. An alarming increase in cataract cases among interventional cardiology (IC) operators has been reported (Vaño et al. 2013). In addition, some case studies of glioblastoma, with an apparent excess in the left (most exposed) hemisphere, have largely contributed to a rising unease (Roguin et al. 2013). While there is, so far, no epidemiological evidence that links brain tumors to occupational exposure, the ICRP recently issued, on the basis of the re-analysis of the cohort of the Hiroshima and Nagasaki bombing survivors, evidence for a new threshold for radiation-induced cataract; it is now thought to be around 0.5 Gy (without dose rate effect) rather than 5 Gy (prolonged exposure). As a consequence, the ICRP recommended lowering the annual eye lens dose limit from 150 mSv to 20 mSv per year (ICRP 2011).

As for the patient, staff dose can be estimated from the $P_{KA}$ delivered to the patient and the protective equipment used by the staff. Since the operator is at variable distances from the patient, the conversion factors are usually difficult to estimate, and are given for a certain distance. The typical order of magnitude of scattered X-ray is around 0.1% to 0.15% of the patient dose at 1 m for a 20 cm $\times$ 20 cm square field (Bushberg et al. 2002).

The ratio between the scattered dose (usually measured in terms of $H^*(10)$) and the PKA is defined as the scatter factor (Schueler et al. 2006; Carnicer et al. 2015). The advantage of using this term is to take into account intraoperative parameters such as patient body habitus, interventional complexity, fluoroscopy time, and acquisition modes. Extensive phantom and in vivo studies have been performed in order to estimate these scatter factors, as well as the main operational parameters influencing these values, as can be seen in Table 66.3.

The main parameters of influence on staff scattered radiation are:

- **Beam angle:** The scattered radiation field has a highly anisotropic geometry. As the number of scattered photons is proportional to the number of incident photons in a given volume, the consequence is a prevalence of

## TABLE 66.3
Scatter Factors Yielded from a Recent Literature Analysis

<table>
<thead>
<tr>
<th>Position</th>
<th>Projection</th>
<th>Scatter Factor [µSv/(Gy cm²)]</th>
<th>Reference</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>Waist (patient plane)</td>
<td>PA 0°</td>
<td>11.0</td>
<td>Schueler et al. (2006)</td>
<td>Baseline: FOV 28 cm, no Cu, patient abdomen thickness 29 cm, radiation field on liver, no equalization filter</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>11.0</td>
<td>Schueler et al. (2006)</td>
<td>FOV 20 cm</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>11.0</td>
<td>Schueler et al. (2006)</td>
<td>FOV 14 cm</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>15.0</td>
<td>Schueler et al. (2006)</td>
<td>+0.2 mm Cu</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>21.0</td>
<td>Schueler et al. (2006)</td>
<td>+0.5 mm Cu</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>15.0</td>
<td>Schueler et al. (2006)</td>
<td>Patient abdomen thickness 24 cm</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>8.2</td>
<td>Schueler et al. (2006)</td>
<td>Patient abdomen thickness 34 cm</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>8.0</td>
<td>Schueler et al. (2006)</td>
<td>Radiation field on spleen</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>9.1</td>
<td>Schueler et al. (2006)</td>
<td>Equalization filter covering 30% of the FOV</td>
</tr>
<tr>
<td>RAO 90°</td>
<td></td>
<td>3.5</td>
<td>Anastasian (2011)</td>
<td>No shield</td>
</tr>
<tr>
<td>NS</td>
<td></td>
<td>1.0</td>
<td>Martin (2009)</td>
<td>Radial access</td>
</tr>
<tr>
<td>NS</td>
<td></td>
<td>1.5</td>
<td>Martin (2009)</td>
<td>Femoral access</td>
</tr>
<tr>
<td>Collar</td>
<td>PA 0°</td>
<td>2.3</td>
<td>Schueler et al. (2006)</td>
<td>Baseline: FOV 28 cm, no Cu, patient abdomen thickness 29 cm, radiation field on liver, no equalization filter</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>2.3</td>
<td>Schueler et al. (2006)</td>
<td>FOV 20 cm</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>3.5</td>
<td>Schueler et al. (2006)</td>
<td>FOV 14 cm</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>5.7</td>
<td>Schueler et al. (2006)</td>
<td>+0.2 mm Cu</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>4.1</td>
<td>Schueler et al. (2006)</td>
<td>+0.5 mm Cu</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>1.1</td>
<td>Schueler et al. (2006)</td>
<td>Patient abdomen thickness 24 cm</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>0.7</td>
<td>Schueler et al. (2006)</td>
<td>Patient abdomen thickness 34 cm</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>2.0</td>
<td>Schueler et al. (2006)</td>
<td>Radiation field on spleen</td>
</tr>
<tr>
<td></td>
<td>PA 0°</td>
<td>2.3</td>
<td>Kuon (2004)</td>
<td>Equalization filter covering 30% of the FOV</td>
</tr>
<tr>
<td></td>
<td>PA 80°</td>
<td>6.6</td>
<td>Kuon (2004)</td>
<td>100 cm from the isocenter (on the right side of the patient, 60 cm adjacent to and 80 cm caudal to the tube)</td>
</tr>
<tr>
<td>Eye</td>
<td>PA 0°</td>
<td>7.0</td>
<td>Vaño et al. (2009)</td>
<td>100 cm from the isocenter (on the right side of the patient, 60 cm adjacent to and 80 cm caudal to the tube)</td>
</tr>
<tr>
<td>Legs</td>
<td>PA, RAO</td>
<td>13</td>
<td>Whitby and Martin (2005)</td>
<td>77 cm from phantom</td>
</tr>
</tbody>
</table>

PA: postero-anterior; RAO, right anterior oblique; NS, not stated.

backscattering around the patient. Beam angles that expose the operator to a larger scatter factor are antero-posterior (AP, X-ray tube above the table) and lateral (with operator at the same side as the tube).

- **Collimation:** Beam collimation does not have an immediate impact on the scatter factor per se (Schueler et al. 2006), but, since both values are proportional, decreasing the field of view (FOV) by using the primary collimation systems reduces the scattered dose, as well as improves image quality, because there are fewer scattered photons impinging on the imaging device.

- **Beam filtering:** In order to spare patient skin dose, low-quality modes usually integrate an extra copper filtration to harden the primary beam. This harder beam increases the mean energy of the scattered radiation, thus increasing the scatter factor (Schueler et al. 2006).

- **Distance to the patient:** As much as the hypothesis of a punctual focal spot holds, this is no longer truly a good approximation for the scatter produced by a patient. Nevertheless, increasing one’s distance to the patient reduces the fluence of scattered photons. Taking a few steps back during a highly irradiating procedure (e.g., DSA) or beam angle helps reduce staff dose. The use of an automatic power injector for contrast administration might be useful, as the operator does not have to stand close to the patient compared to manual (syringe) contrast medium injection. The access site for percutaneous catheterization interventions also plays a major role in staff exposure, as they determine the proximity of the operator to the primary field. Generally, for trunk examinations, the scatter factors are higher for radial approaches than for femoral approaches. Jugular approaches (e.g., TIPS or myocardial biopsy) yield even higher factors due to field proximity and the usual lack of space to properly use leaded shielding.

- **Shielding:** In addition to personal protective equipment (lead apron and thyroid collars), several other means of protection are available, including (but not limited to):
  - **Leaded glasses:** Typically, the lead equivalence of protective glasses is 0.5 to 0.75 mmPb. A more covering but less protective alternative (0.1 mmPb) are protective face shields, whose geometry allows for a simultaneous protection of the operator’s encephalon.
  - **Leaded gloves:** These surgical gloves are made in the same way as other current lead garments, such as a metal dispersion (bismuth, antimony, or barium as a surrogate to lead) in a polymer matrix. Their efficiency is usually less than 0.5 mmPb (McCaffrey et al. 2012). The downside of these gloves is a false sense of security that could lead some physicians to leave their hands in the primary beam. This can potentially lead to high operator extremity doses—as these protective devices are solely made to attenuate scattered (i.e., less energetic) radiation. Furthermore, their attenuation power may lead to an increased radiation burden on the patient as the AEC system responds to a denser object in the X-ray field.
  - **Protective patient drapes:** A recent innovation is a sterile draping of several sizes, depending on the purpose, made of an antimony-bismuth dispersion in a polymer matrix. This protection is to be placed upon the patient, outside the primary beam—to avoid the effects mentioned for the leaded gloves—in areas where scatter radiation leaves the patient. The purpose is to form a “shadow” for the operator standing next to the patient. The efficiency is not quite known, and may highly depend on the other protective devices used alongside the protective drapes. Its sterile cover makes it single-use and a rather costly solution.
  - **Structural shielding:** Several shielding devices can be permanently mounted in an angiography room. Most common solutions include table-mounted leaded curtains (some with movable parts), ceiling-suspended leaded acrylic glass, and floor-mounted movable leaded acrylic panels. The aim of these protections is to be constantly adapted to the irradiation geometry and the patient’s body. This protective shielding can be improved by hanging small leaded curtains on the parts where a gap may be present between the patient and the panel.

Figure 66.8 summarizes the previously mentioned means of protection in their typical clinical position.

The efficiency factors of these means of protection are defined as the ratios of the doses with and without the protection. For dose calculation and optimization purposes, the ideal situation would be to dispose of a single efficiency factor for each means of protection, which could be multiplied together to estimate the
Global efficiency of a given set of protections. Unfortunately, the efficiency of these different means of protection is highly dependent on their position with respect to the patient and the operator. Furthermore, phantom measurements or Monte Carlo simulations for these means of protection usually yield higher efficiency factors than in vivo measurements performed during surgery (Martin 2010). For example, the efficiency factors mentioned for the thyroid collar can reach 5–6 for Monte Carlo simulations, 4–6 for in vivo simulations, but only 2–3 for staff measurements. Furthermore, the efficiency factor for leaded glasses is stated at around 30 using practical simulations (beam facing the glass plane), but this efficiency could drop to a mere 1.25 for lateral angles and when the operator is looking away from the field (e.g., to a ceiling-suspended diagnostic screen) due to grazing incidence (Koukorava et al. 2014).

The inherent uncertainties linked to retrospective and/or prospective staff dose estimations imply that currently the only effective solution still relies on dosimetry, using double dosimetry for effective dose and surface dose estimation, and extremity dosimetry for finger dose measurement. Furthermore, a new dosimetry value has been proposed by several authors (Martin 2016; IRSN 2014b), namely 

\[ \text{Hp}(3) \]

for example, an absorbed dose at 3 mm depth, intermediate between 

\[ \text{Hp}(0.07) \]

and 

\[ \text{Hp}(10) \]

using a dedicated dosimeter placed on the physician’s head.

### 66.4 Radiation Protection in Computed Tomography

#### 66.4.1 Technical Requirements

Computed tomography has changed the practice of medicine. From its range of clinical applications and outcomes it has enhanced the understanding of diseases and improved patient care (Rubin 2014). However, CT remains a high dose procedure, and its optimization remains a priority.

As for radiography or fluoroscopy equipment, the first step of radiation protection is to ensure that the CT unit complies with standardized safety requirements such as the ones published by the IEC (2004, 2009b). As for all radiological devices, constancy tests should be applied in the framework of a quality assurance program. An IEC standard also defines the main content of such a process (IEC 2006). The full characterization of a CT unit is also extensively described in a special ICRU issue dedicated to CT imaging (ICRU 2012). This contribution will not describe in detail the methods that can be used to characterize a CT unit, but will concentrate on the main items concerning the assessment of image quality. In the framework of quality control there are several test objects, such as the Catphan® phantom shown in Figure 66.9, that are available to evaluate the overall quality of the images produced by the unit.

To assess the stability of the Hounsfield numbers and check for the reconstructed slice thickness and absence of geometric distortion, one can use Figure 66.9a. The other slices make it possible to, respectively, assess the spatial resolution (Figure 66.9b), the low contrast resolution (Figure 66.9c), and the image homogeneity (Figure 66.9d). To facilitate the assessment of these parameters, computer programs have been developed by several groups (Pahna et al. 2016). These various methodologies assess image quality partially at Level Ia according to Table 66.1, since the efficiency of the CT unit, assessed by a quantity such as a DQE type, remains quite difficult to implement. A few years ago, several groups proposed quality factors to characterize CT units. However, these quantities were based on signal theory that required signal linearity, and were not task-based oriented which limited their use for practical CT optimization. They were employed mainly for unit benchmarking (Brooks and Di Chiyo 1976; Riederer et al. 1978; ImPACT 1998; Nagel 2012).

Over recent years, manufacturers have developed strategies and devices to reduce patient exposure as much as possible while providing a level of image quality compatible with the diagnostic requirements. Among the major technological advances worth mentioning is the introduction of the automated tube current modulation (ATCM) (Kaza et al. 2014). This is of major importance, since, unlike plain analog film radiography, where excessively high radiation exposures result in reduced image quality or even a non-readable image, digital radiography or CT image quality does not deteriorate at high exposures. On the contrary, images become even more comfortable to read due to the reduction of the noise level. Some tube current modulations (i.e., organ-based tube current modulation, OBTCM) can even take into account the fact that, during the acquisition, the X-ray will expose particularly radiosensitive organs such as the eye lenses, the thyroid, or the breast. However, if the idea seems rather enticing, the optimal positioning of the sensitive organs remains a major issue in order to fully benefit from the technique (Taylor et al. 2015). Recently, X-ray tube high voltage modulation (kV modulation) has been proposed as one strategy for further reducing the radiation dose when dealing with imaging of high Z materials, taking advantage of the benefit of the photoelectric effect (Wintermark et al. 2000; Suh et al. 2013). Much progress has also been made concerning not only the efficiency of the detector, but on the penumbra of the beam. Some advances in terms of the X-ray beam collimation have also been proposed, such as dynamic asymmetrical beam collimation to reduce unused patient exposure at both ends of the planned acquisition volume during helical scanning (so called: adaptive collimation) (Deak et al. 2009).

The way information is extracted from the data has also been drastically improved by the introduction of iterative reconstruction. Users are able to produce readable images at very low dose levels (Den Harder et al. 2015; Naoun et al. 2015; Padole et al. 2015). However, image content remains, up to a certain level, dose dependent, and one should remember that drastic dose reduction impairs, in particular, the detection of low contrast structures (Schindera et al. 2013; Ott et al. 2017). Finally, if a CT unit, when it complies with international standards, is a relatively safe piece of equipment, the way it is used on patients might lead to undesired effects (Imanishi et al. 2005; Rehani 2015a,b). To limit overexposure as much as possible, some manufacturers have proposed software that checks whether the settings of the protocol could lead to acute tissue reactions or present major errors (NEMA 2010; Howard et al. 2014; Miller et al. 2014). On top of this, manufacturers offer training sessions to ensure that all the technological solutions available on the unit are optimally used (EURATOM 2013; IAEA 2014).
66.4.2 Optimal Way to Use the Unit

Similar to radiography or fluoroscopy, the next step of the optimization process should be done with the clinical applications in mind. The assessment of the unit for generic protocols is generally performed by medical physicists who should be involved in setting quality assurance programs (EURATOM 2013; IAEA 2014). Level Ia, as defined in Table 66.1, can be made with standard test objects where image quality criteria will often be far from clinically relevant tasks (Verdun et al. 2015). When willing to optimize actual clinical protocols, as for radiography, we should use task-based methodologies. Level Iia can be reached by using mathematical model observers and standard quality assurance phantoms, as shown by several groups (Hernandez-Giron et al. 2011, 2014; Koffler et al. 2015; Samei and Richard 2015). To reach Level Iib, one can use anthropomorphic phantoms and either human or mathematical model observers (Yu et al. 2013; Tseng et al. 2014; Zhang et al. 2014; Ott et al. 2017; Racine et al. 2016). If these approaches introduce a task, thus narrowing the gap between Levels Ia,b—which consider metrics whose outcomes are difficult to use in a clinical environment—and Level III, they still lack in anatomical realism. The phantoms used are generally homogeneous, and can be optimized with a task that remains far from clinical situations, overestimating, for example, the potential dose reduction when using iterative reconstruction (De Crop et al. 2015). The development of 3D printing technology will certainly be very useful in the future to produce images where the background is more tissue-like structured (Solomon and Samei 2014). Level III methods of assessing the adequacy of image quality with the diagnostic task have been used extensively, with the introduction of iterative reconstruction trying to lower patient exposure as much as possible while keeping the diagnostic required information (Miéville et al. 2013; Smedby et al. 2013; Zarb et al. 2015). As mentioned previously, in spite of their difficulty to implant, they remain necessary when drastic changes of dose and data processing are introduced.

Having briefly described patient dose and image quality while considering image quality at the first stage, other initiatives have used patient dose instead. Dose quantities are a lot easier to assess than image quality, which should be related to diagnostic tasks. The introduction of DRLs has certainly helped to reduce the spread of patient exposure, but these values are still poorly related to a specific diagnostic task. They are often considered to be limits, which is not the case, and concepts aiming at reducing patient dose as much as possible, such as the introduction of “target dose” values or “achievable dose,” might be counterproductive in the future. Image quality should remain a main priority of the optimization process (Rehani 2015a,b). At the same time,
tailoring image quality to diagnostic requirements increases the number of CT protocols. If this process is important in the context of radiation protection, it introduces a serious concern because of the lack of standardization. One of the first efforts to standardize CT acquisitions was applied quite successfully in the pediatric population by color-coding the protocols, but there is still a way to go (Frush et al. 2002; Singh and Kalra 2014). Nevertheless, many initiatives are focused on protocol standardization meant to enable an optimal use of the software solutions and making it possible to compare between centers in terms of patient exposure (Kofler et al. 2014; Escalon et al. 2015; MacGregor et al. 2015). This is certainly a quality criterion, but image information should not be forgotten. Finally, as for radiography, several websites are available to help the radiographers and radiologists to optimize their clinical protocols. The ones already mentioned for radiography (Image gently and Image wisely) have a specific section for CT protocols (IG 2015; IW 2015). At the European level, the European Society for Radiology has published a white paper of the radioprotection aspects in CT (ESR 2011) and provides also a website to help the CT users in the optimization process of their protocols (ESR 2016).

CT units can be used for 4D perfusion studies where local dose levels can be quite high. To avoid overexposures, such as those mentioned at the beginning of this chapter (Imanishi et al. 2005; FDA 2010; Coeytaux et al. 2015), the users should monitor the cumulated CTDIvol indicated on the unit before the image acquisition.

CT units are also used for guiding therapeutic procedures (interventional radiology), whose level of complexity might also require relatively long exposition times. As for perfusion studies, the monitoring of the cumulated CTDIvol is of primary importance to inform the patient if exposure is over the threshold of the deterministic risks (Leng et al. 2011).

Finally, optimization in radiology should not only consider the radiation risks, but other risks such as the one associated with the use of contrast media (Noda et al. 2015). Drastic dose reductions or beam filtering might reduce the contrast-to-noise ratio so that the amount of contrast media needs to be significantly increased. Radiology is a matter of compromise.

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Radiation Protection Issues in X-ray Radiology, Fluoroscopy, and Computed Tomography


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