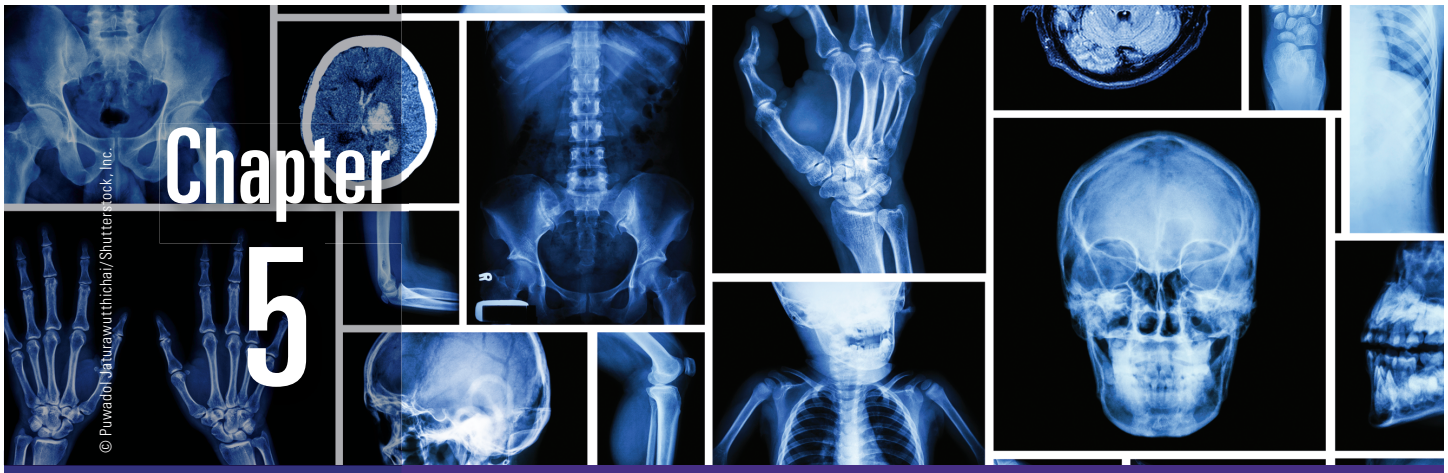

Chapter 5

Radiation Protection Practice

Outline

- Introduction
- Dose Risks in Diagnostic Imaging
- Justification
  - What is justification?
  - How is justification implemented?
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  - What is optimization?
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  - Thermoluminescent dosimetry (TLDs)
  - Dose-area product meters (DAPs)
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- References

Objectives

On completion of this chapter, you should be able to:

1. Identify the cancer risks for common, non-fluoroscopic and non-CT examinations.
2. State and discuss two well-known principles have been adopted within legislative, regulatory, and advisory documents to promote keeping the risk to patients as low as reasonably achievable.
3. Discuss the essential considerations of the principles of justification and optimization.
4. Explain briefly what is meant by the term "non-optimization."
5. Identify new technologies that play a role in dose optimization.
6. Discuss a number of optimization initiatives.
7. Differentiate between *image quality* and *diagnostic efficacy* in the optimization process.
8. Explain what is meant by dose limits.
9. State the dose limits for occupational individuals and members of the public.
10. Identify and discuss four methods of radiation detection and measurement.

Introduction

Every radiation exposure in medical imaging departments will introduce a risk of inducing a cancer, but it should present a benefit to the patient as long as the exposure is justified. The responsibility of the radiographer or radiologic clinician is to ensure that the radiation dose is minimized and the benefit maximized for each examination that takes place. The "as low as reasonably achievable" (ALARA) principle should be applied each time we expose the patient, meaning that the radiation risk is kept to a low level, but not to the extent where the diagnostic efficacy of the examination is compromised—and techniques and technologies have been developed around this objective.

But there is more: Systems and policies must also in place to ensure that the radiation dose to individuals who are *not* patients, but who nonetheless may be exposed as a result of diagnostic examinations in their workplace, is also kept as low as possible. Such individuals include healthcare workers, people transporting patients (e.g., first responders, family members, etc.), and other people in the vicinity at the time of the exposure. In this context, radiation protection standards have evolved.

Three main systems are in place to maintain good protective standards: (1) *justification* to make sure that there is a good reason for performing each x-ray examination; (2) *optimization,* which will promote the best possible methods and technologies for imaging the patient at the lowest dose; and (3) *dose limit*s for those other individuals who may inadvertently (and without any direct benefit) undergo radiation exposures during patient examinations. With regard to dose limits, it is very important to highlight that these limits or dose constraints *do not apply to the patient*; if the examination dose is justified, then that dose is delivered. Admittedly, we do have diagnostic reference levels for patients to offer guidance levels for doses that should be administered, but these are not dose limits (**Figure 5-1**).

The three principles of justification, optimization, and dose limits are the focus of a number of recent legislative, regulatory, and advisory documents addressing radiation protection. For example, in the European Council Directive 97/43/EURATOM *On Health Protection of Individuals Against the Dangers of Ionizing Radiation in Relation to Medical Exposure and Repealing Directive 84/466/Euratom* (ECD 1997; Teunan 1998), which has been adopted into national legislation in a number of European states, articles 3 and 4 are dedicated (respectively) to justification and optimization. In the European Council Directive 96/29/Euratom *Laying Down Basic Safety Standards for the Protection of the Health of Workers and the General Public Against the Dangers Arising from Ionising Radiation* (1996), again incorporated within national legislation across Europe, Chapter I of Title IV deals with justification and optimization, whereas Chapter II of the same Title focuses on describing specific dose limits. These and other legislative or advisory documents will be discussed throughout this chapter (ARPANSA 2008; ICRP 2007).

Finally, a fundamental component of good radiation protection standards is ***effective dosimetry***; i.e., methods of measuring radiation doses to patients, workers and other exposed individuals. These methods range from personal dosimetry devices, which are worn at all times by individuals working with radiation, to the more sophisticated electronic dosimeters—a fundamental tool for quality assurance procedures ensuring radiation protection standards. Dosimetric methods will be considered toward the end of this chapter.

Dose Risks in Diagnostic Imaging

Before appropriate radiation protection methods can be described, we need to put into context the risks that are associated with medical imaging. As has been discussed elsewhere, the risk of inducing a cancer using doses typically encountered in diagnostic imaging departments is low. The majority of non-fluoroscopic and non-CT examinations involve effective doses somewhere between 0.0001 mSv for a dorsiplantar projection of the foot, up to 0.5 mSv for an anteroposterior (AP) projection of the abdomen. These doses for a 20-year-old

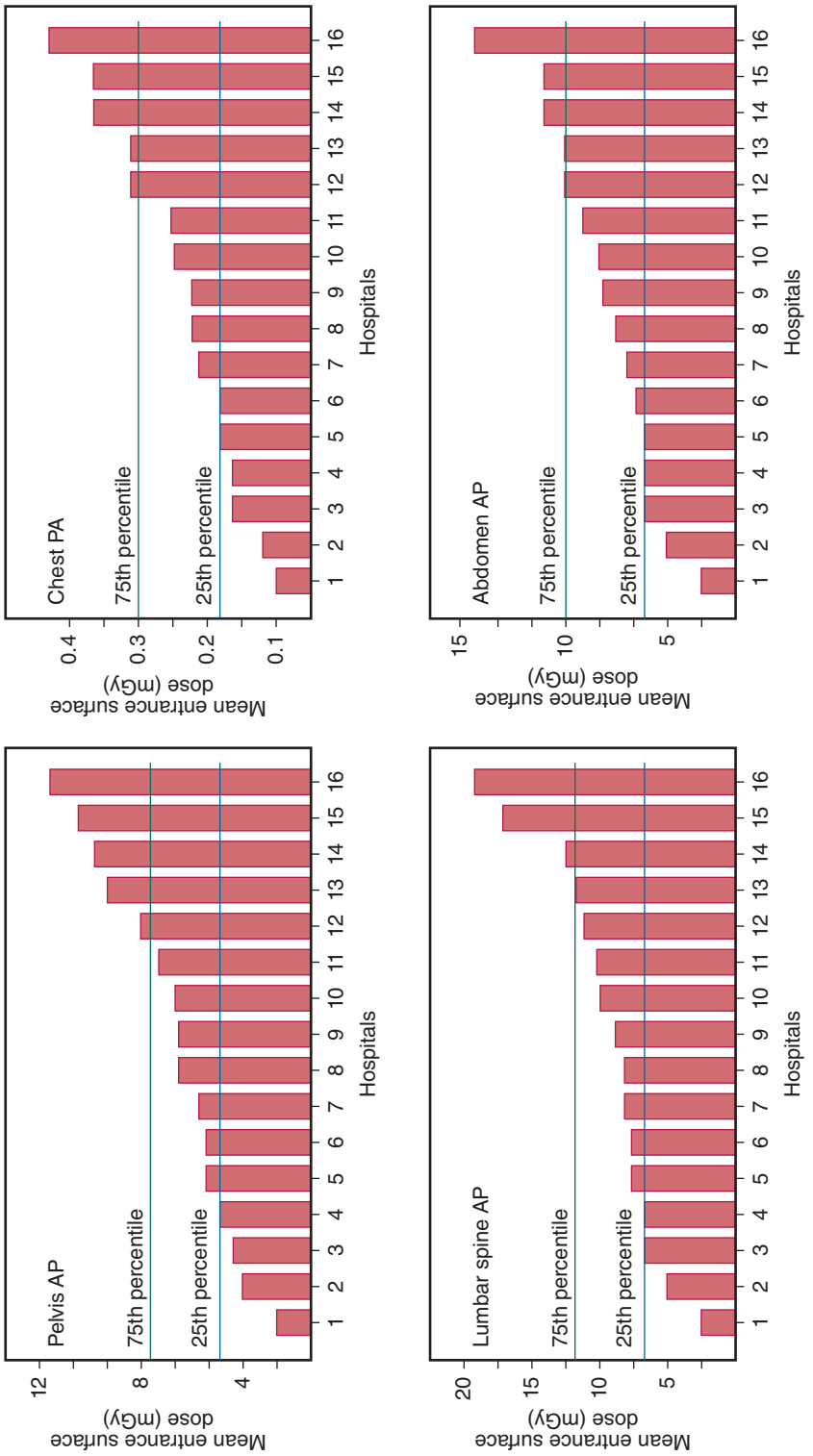


Figure 5-1 The 75th percentile demonstrates DRL values for a number of examinations.

Data from: Johnston DA, Brennan PC. Br J Radiol. 2000 Apr;73(868):396–402.

individual equate to a fatal cancer risk of approximately 1 per billion and 3.5 per 100,000, respectively (yet we must still remember to take great care in assessing individual risks). A summary of effective doses and cancer risks for these types of examinations are provided in **Tables 5-1** and **5-2**), respectively (Wall et al. 2011).

Table 5-1 A summary of effective doses for common non-fluoroscopic and non-CT examinations

Radiograph	103 (mSv)
Abdomen AP	0.43
Both hips AP	0.19
Cervical spine AP	0.018
Cervical spine Lat	0.012
Chest Lat	0.038
Chest PA	0.014
Femur AP	0.011
Femur Lat	0.001
Foot (dorsi-plantar)	0.0001
Foot (olique)	0.0001
Head AP	0.033
Head Lat	0.016
Head PA	0.02
Knee AP	0.0001
Knee Lat	0.0001
Lumbar spine AP	0.39
Lumbar spine Lat	0.21
Lumbo-sacral joint Lat	0.17
Pelvis AP	0.28
Shoulder (Axial)	0.004
Shoulder AP	0.007
Single hip AP	0.087
Thoracic spine AP	0.24
Thoracic spine Lat	0.14

AP = Antero - posterior

PA = Postero - anterior

Lat = Lateral (average of left and right lateral)

Of course, there is the issue of higher doses being received from interventional investigations and CT examinations, notwithstanding the associated deterministic effects that may occur once specific dose thresholds are exceeded (which are still not commonplace). These higher doses can present a risk that can be considerably higher; for example, risk of inducing a cancer to a young child undergoing CT of the whole trunk might be in excess of 1 in 1000. Other high-dose examinations can deliver typical effective doses; coronary angiography and whole-trunk CT can produce doses of 3.9 mSv and 10 mSv, respectively, with associated cancer risks in a 50-year-old male of 1.9 and 3.5 per 10,000. A more complete description of effective doses and cancer risks for these higher dose examinations are given in **Tables 5-3** and **5-4** respectively.

You will see from these tables that the lifetime risk of cancer varies significantly between ages and gender. The level of variation depends on the examination type. This highlights the caution that one must apply when employing parameters such as effective dose and detriment-weighted nominal risk coefficients, which, averages the parameters over genders, ages, and indeed different populations in an effort to provide single values that clinicians can work with more easily.

Presenting such risks in a meaningful way to patients or even the general population is not easy, since the concept of risk is a difficult one. If we take the cancer risk following an AP abdominal exposure, estimated at roughly 1 in 30,000 (3.5 in 100,000), it may be worth comparing this with the lifetime risk to the US population of dying from heart disease (1 in 5), cancer (1 in 7), falling down (1 in 246), or electrocution (1 in 5000). In fact, the chance of inducing a cancer following an AP abdominal exposure is similar to that of dying in a flood. I will leave it to the reader to decide whether these comparisons are of any value.

In summary, it is clear that, apart from a minority of examinations such as those involving CT and interventional procedures, the risk is low for most examinations. The question must be

Table 5-2 A summary of cancer risks for common non-fluoroscopic and non-CT examinations

Examination	Sex	Age at exposure (y)									
		0–9	10–19	20–29	30–39	40–49	50–59	60–69	70–79	80–89	90–99
Head	M	12	8.5	5.9	4.4	3.2	2.2	1.3	0.6	0.3	0
(AP+PA+Lat)	F	11	7.7	5.3	3.7	2.9	1.8	1	0.5	0.2	0
Cervical spine	M	2.7	1.9	1.2	0.9	0.6	0.4	0.2	0.1	0.1	0
(AP+Lat)	F	6.2	3.7	2.2	1.3	0.8	0.5	0.3	0.2	0.1	0
Chest	M	1.3	1.1	0.9	0.8	0.7	0.6	0.5	0.3	0.1	0
(PA)	F	1.9	1.6	1.4	1.3	1.2	1.1	0.8	0.5	0.2	0
Thoracic spine	M	30	24	20	17	16	13	9.7	6.1	2.6	0.1
(AP+Lat)	F	65	50	40	34	30	25	18	11	4.2	0.1
Abdomen	M	55	44	35	27	21	15	9.3	4.8	1.7	0.1
(AP)	F	49	39	31	25	20	14	9.4	5.2	1.8	0
Pelvis	M	31	25	20	16	13	9.4	5.9	3	1	0
(AP)	F	24	19	16	13	10	7.8	5.2	2.9	1	0
Lumbar spine	M	72	56	43	34	26	19	12	6.1	2.3	0.1
(AP+Lat)	F	65	51	41	32	26	19	12	6.8	2.4	0.1
Knee	M	0.011	0.008	0.005	0.004	0.003	0.002	0.001	0	0	0
(AP+Lat)	F	0.008	0.006	0.004	0.003	0.002	0.001	0.001	0	0	0
Foot	M	0.0049	0.0035	0.0024	0.0017	0.0012	0.0007	0.0004	0.0002	0	0
(AP+Lat)	F	0.0036	0.0026	0.0019	0.0013	0.0009	0.0006	0.0003	0.0002	0	0

Table 5-3 A summary of effective doses for fluoroscopic and CT examinations

Examination	103 (mSv)
Coronary angiography	3.9
Femoral angiography	2.3
CT Abdomen	5.6
CT Abdomen + Pelvis	6.7
CT Chest	6.6
CT Chest + Abdomen + Pelvis	10

asked, however: If the risk is already low, is it still important or worth the effort to further protect the patient (or staff) by having radiation protection standards? While we will look at this question in more depth as this chapter progresses, the simple response is to ask ourselves that if we *can* produce images of equal diagnostic efficacy at lower risks to the patient (even if that risk is already very low), as radiographers and clinicians, do we not have that responsibility to the patient? Particularly if we can reduce (or even remove) exposures at minimum (or no) extra cost or effort?

Two well-known principles have been adopted within legislative, regulatory, and advisory documents to promote keeping the risk to patients as low as reasonably achievable: *Justification* and *Optimization*. Each of these principles will now be considered.

Justification

What is Justification?

Each x-ray exposure must have a good medical reason for performing it. Irradiating individuals introduces a risk, yet this risk is acceptable if it is outweighed by the benefit that is provided diagnostically by the resultant image or images. In other words, there must be good *justification* regarding the exposure; in the words of the International Commission on Radiological Protection (ICRP 2007), it should "*do more good than harm for the patient.*" The whole issue of justification has been well argued by Matthews et al. (2008), which will be referenced in this section.

First, let us look at how legislation or guidance documents define justification. In the European Council Directive 97/43/EURATOM, Article 3 (ECD 1997; Teunan 1998) defines justification thus:

> *Medical exposure … shall show a sufficient net benefit, weighing the total potential diagnostic or therapeutic benefits it produces, including the direct health benefits to an individual and the benefits to society, against the individual detriment that the exposure might cause, taking into account the efficacy, benefits and risks of available alternative techniques having the same objective but involving no or less exposure to ionizing radiation.*

The same document makes it quite clear that if the exposure cannot be justified, then it should be prohibited.

In its *Initiative to Reduce Unnecessary Radiation Exposure from Medical Imaging* (FDA 2012), the Food and Drug Administration within the United States presents a similar definition of justification:

> *The imaging procedure should be judged to do more good than harm (e.g., detriment associated with radiation induced cancer or tissue effects) to the individual patient. Therefore, all examinations using ionizing radiation should be performed only when necessary to answer a medical question, treat a disease, or guide a procedure. The clinical indication and patient medical history should be carefully considered before referring a patient for any x-ray examination.*

Other documents from a variety of other countries, including Australia in its *Radiation Protection in the Medical Applications of Ionizing Radiation* (ARPNSA 2008), similarly highlight the importance of justification (and optimization).

Table 5-4 A summary of cancer risks for fluoroscopic and CT examinations

Examination	Sex	Age at exposure (y)									
		0–9	10–19	20–29	30–39	40–49	50–59	60–69	70–79	80–89	90–99
Coronary angiography	M	330	290	250	230	210	190	150	94	41	2.1
	F	430	390	370	360	370	330	270	170	66	1.7
Femoral angiography	M	280	220	170	140	110	85	56	32	14	1.6
	F	210	170	140	110	110	73	45	24	8.8	0.5
CT head	M	250	190	130	100	80	57	36	20	9	1.2
	F	190	140	100	77	71	46	27		4.8	0.3
CT chest	M	530	440	350	300	260	220	160	99	42	2.2
	F	1100	860	680	560	490	390	290	180	68	1.7
CT abdomen	M	670	530	400	310	240	170	110	56	21	1.5
	F	610	480	380	300	240	170	110	59	20	0.6
CT abdomen + pelvis	M	850	670	520	410	320	230	150	78	29	1.9
	F	740	590	470	370	310	230	150	80	28	0.8
CT chest + abdomen + pelvis	M	960	780	630	520	440	340	240	140	58	3.3
	F	1500	1100	910	740	640	500	260	210	80	2.1

It should be stressed that a number of the well-known perspectives on justification rely on the details contained within the ICRP 103 (ICRP 2007) recommendations.

How is Justification Implemented?

Justification is implemented at two main levels: First, at a broad level, the use of a specific examination or a specific imaging tool must be considered justifiable by an industry and/or governmental accreditation board before it can be adopted by health services, screening programs, imaging departments, insurance providers, and so on. The second point of justification is specific to each patient: Before any individual exposure takes place, the radiographer or clinician must be satisfied that there is a good medical benefit for providing such an exposure. Both broad and specific implementation will be considered here. Of course, it can be argued (as in ICRP [2007] and Matthews et al. [2008]) that there is a more fundamental level of justification, which is that the introduction of x-rays at all must demonstrate a net benefit compared with any detriment.

Broad Implementation of Justification

In the European legislation (1997), justification at the broader level should be evident (1) whenever *a new or alternative method* of method imaging that uses ionizing radiation is being proposed, and (2) for *all existing methods when new data regarding efficacy has been made available.*

When a new or alternative method of imaging is being proposed, it must be carefully evaluated to make sure that an overall benefit will accrue to the population on whom this technology or technique will be used. A modern example may help explain this: In most countries (**Figure 5-2**), breast cancer screening uses mammography as the first-line tool via 2-dimensional cranio-caudal (CC) and medio-lateral oblique (MLO) images taken of each breast (**Figures 5-3**).

Over the last few years, there has been increasing evidence that a new modality—digital breast tomosynthesis (DBT), which presents the images in discrete slices or as 3-dimensional images (Alakhras et al. 2013; **Figure 5-4**) may offer important benefits in terms of increased cancer detection rates and lower numbers of false positives (e.g., cases where an anomaly is identified by the radiologist as a cancer that later proved to be normal or benign tissue). A study by Skaane et al. (2013) involving over 12,000 women whose breast images were interpreted using mammography alone or mammography combined with DBT concluded that there were significant benefits when the new approach was used with this population of women.

Figure 5-2 Typical mammographic equipment used in breast screen imaging.

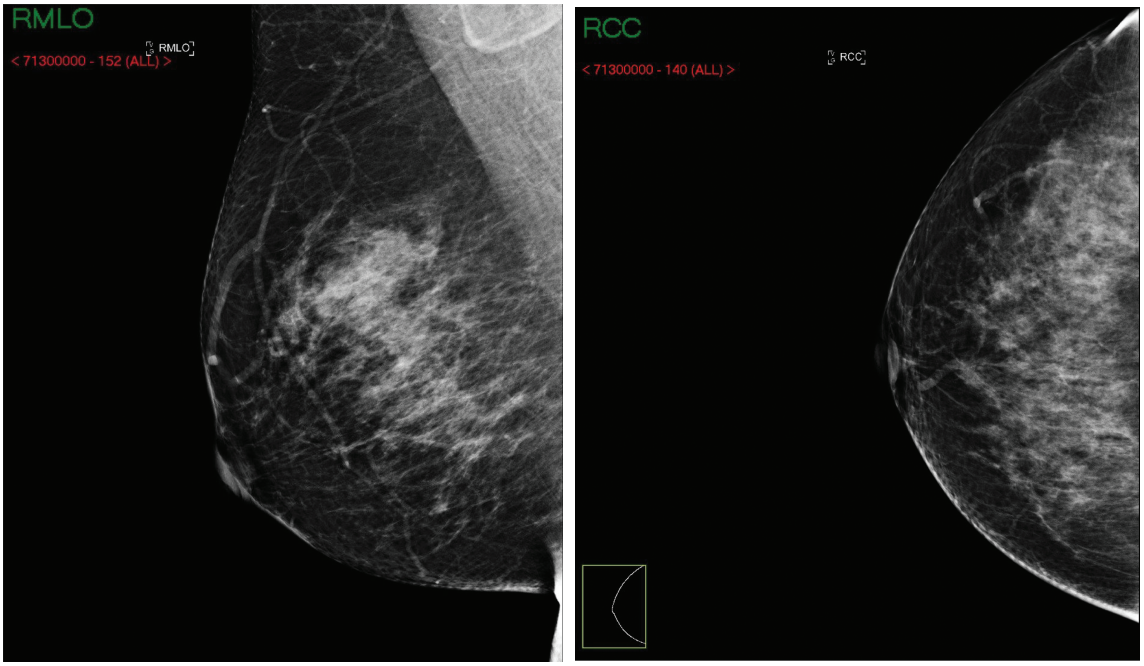


Figure 5-3 CC (A) and MLO (B) images typically produced for each woman presenting at a breast screening clinic.

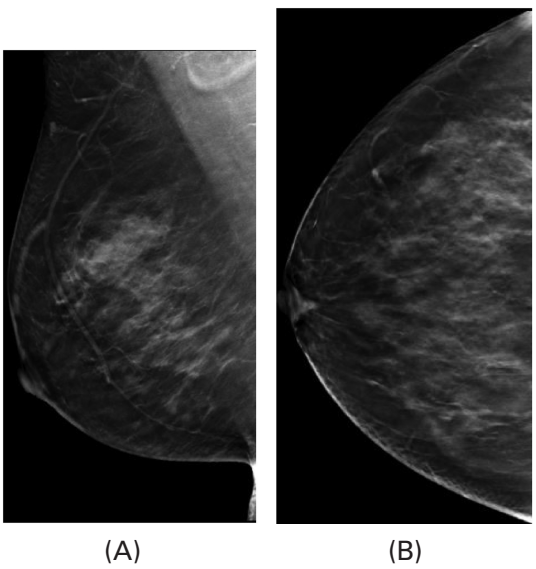


Figure 5-4 (A) MLO projection of the breast (digital). (B) CC projection of the breast.

While these results and others looking at DBT clearly offer a good basis for a justifying this new technology, more data are required before a full justification to replace mammography is presented. These data would include:

- Full radiation dose assessment so any increased risk to women being examined is understood
- Impact on radiographic and radiologic work practices, since longer examination times may result in fewer women being x-rayed
- Full health economic assessment, to make sure that a specific population's financial situation can support implementation of this new tool
- Whether the new tool will actually replace the old or serve as an adjunct technology

It is also important to make sure that evidence is relevant to the population to which a new technology or technique is being introduced. Again using

the mammography example, while there is encouraging data on DBT coming out of the United States, Europe, and Australia, will this be useful to women in the Middle East and Southeast Asia, for example, where the types of cancer and the age profile of women with breast cancer can be very different from those in the Western world? It is often critically important that individual states or countries introduce their own evidence to support a change in practice or equipment when their circumstances are different, rather than simply (and conveniently) relying on data that has been produced from elsewhere.

While rigorous studies are usually performed to explore whether a novel practice or technology should be implemented, often there is less emphasis on justifying current activities. Let us remind ourselves again that, according to at least one set of legislation, "existing types of practices involving medical exposure may be reviewed whenever new, important evidence about their efficacy or consequences is acquired." This means that if a new, scientifically robust study demonstrates that the *current* method of x-raying an abdomen in the AP position is not optimal and might be improved if a mobile patient was placed in the posteroanterior (PA) position, then the current practice of using AP positioning should be reviewed. Unfortunately, in the experience of this author, review of common radiographic procedures rarely takes place, even though evidence does emerge from time to time of potentially better ways of x-raying our patients. This discussion does cross over to optimization of techniques, and the argument of needing to have systems in place to revisit long-established procedures will continue. A good example of an effective review system that covers UK health delivery much more broadly is the one implemented by the National Institute for Health and Care Excellence (NICE), where specific topics are referred to NICE for consideration; these are effectively reviewed, and recommendations for best practice are provided based on the latest evidence. It is difficult to see how current radiographic practices can be thoroughly justifiable without a similar, albeit more focused, system of review. With the risks associated with radiographic exposures setting up an effective review system should be, in the opinion of the author, an issue of priority for international, national, and local professional and regulatory bodies.

Individual Implementation of Justification

According to the European legislation (ECD 1997; Teunan 1998), "all individual medical exposures shall be justified in advance, taking into account the specific objectives of the exposure and the characteristics of the individual involved." This level of justification occurs, therefore, each time we examine a patient radiologically, and it means that before any exposure is performed, a good clinical reason for that exposure must be provided. To facilitate justification, the European legislation is clear that all relevant prior details, such as previous images along with the relevant medical records, should be available and considered so that unnecessary exposures are avoided.

So who is responsible for making sure that each individual exposure is justified? While this responsibility may vary from jurisdiction to jurisdiction, at least in a number of European states, this role appears to fall into two categories: The person requesting the exposure and the person with the responsibility to perform the examination. The person requesting the exposure must provide sufficient clinical reason for the procedure. In the UK and Ireland in the early 2000s, to make sure that referrers such as general practitioners (GPs) and hospital-based medical doctors were equipped with the necessary information to allow an appropriate referral, each received a set of referral guidelines (Royal College of Radiologists 2003). These guidelines have been subsequently updated (Royal College of Radiologists 2007).

While in the United States, according to the FDA, the referrer has the primary responsibility for justifying the examination, in Europe the second category of persons—those with responsibility for delivering the examination—is also

critically important to the individual justification process. This importance was highlighted when it was shown in three European member states that significant numbers of medical exposures for common radiographic examinations did not have sufficient referral justification (Bell and McLaughlin 2001; Triantopoulou et al. 2005; Morris-Stiff et al. 2006). The European legislation therefore specifically refers to an individual described as a *practitioner* who will be involved in the justification process, that practitioner being defined as a "medical doctor, dentist, or other health professional who is entitled to take clinical responsibility for an individual medical exposure." It has been debated as to whether the radiographer or radiologic technologist is in the position to be either the practitioner or someone who acts as an interface between the prescriber and the practitioner, particularly when the clinical information provided by the prescriber is insufficient. Whatever the official role might be—and this will vary from one locality to another—it is clear that the person who is responsible for delivering the radiation, most likely the radiographer or technologist, will contribute importantly to the justification process since, ultimately, according to 97/43/EURATOM (ECD 1997; Teunan 1998), "if an exposure cannot be justified it should be prohibited."

The emphasis above has focused on demonstrating the need for clinical justification before any exposure is performed. However, there are two important exceptions to this clinical justification. The first involves the provision of x-ray images, particularly of the chest, for immigration, employment, or medico-legal purposes. In such instances, we do not have a good *clinical* justification for each individual exposure, but nonetheless these are performed on a reasonably regular basis. The European legislation simply says that justification for these procedures is the need for "special attention" (1997). The second exception is when x-ray exposures are performed for research purposes; however, these exposures should be (and usually are) approved by rigorous ethical application process and restricted to the specific protocols approved by the ethics committee in line with local and national policy.

Optimization

What is Optimization?

Optimization of x-ray procedures means employing technologies or techniques that can reduce the radiation dose to patients (and staff) while not sacrificing in any way the clinical information (relevant to the patient's condition) provided by performing the examination. European Council Directive 97/43/EURATOM (ECD 1997; Teunan 1998) describes optimization thus:

> All doses due to medical exposure for radiological purposes…Shall be kept as low as reasonably achievable consistent with obtaining the required diagnostic information taking into account economic and social factors.

The question was posed earlier: If the risk to patients from an array of medical x-ray procedures is low, is it worth us putting time and effort into making sure that we present an even lower risk by optimizing exposures? A second question was posed as well: If producing images of equal diagnostic efficacy at lower risks to the patient is possible, do radiographers and clinicians have a duty or responsibility to the patient to keep those risks to a minimum, particularly if lower dose measures are not costly or inconvenient? To the author, this latter proposition is a good enough reason to lower the doses wherever possible, but other arguments are presented now, starting with a real and modern-day context.

One of the most frequently discussed examinations when it comes to benefit versus risk is the breast cancer screening mammogram. The value of this examination is hotly debated, with the majority of the evidence coming down on the side of supporting breast cancer screening strategies. However, there is a persistent anti-screening argument: In addition to the issue of overdiagnosis, the other

often-quoted concern associated with x-raying well women is the radiation dose delivered during the screening with its associated risk of inducing a cancer. Radiation protection practices are particularly tightly controlled in breast screening, and subsequently the typical mean glandular dose (MGD) for a mammogram is around 3–5 mSv. According to the nominal risk coefficient stated within ICRP 103 (ICRP 2007), this would mean that in a cohort of 1 million women, each receiving the above dose, the risk would be about 30 to 50 induced breast cancers *per million examinations*—a number increasing by a factor of between 7 and 10 over a set period in which women receive repeated screenings (say, 20 years). This should be balanced against the estimated 5000 breast cancer deaths prevented per 1 million women by the breast screening program (Marmot et al. 2012). Would it make much difference if the radiation protection standards were not as tight, and the resultant doses increased by, say, a factor of 3 or 4 (which is the typical variation seen across centers for other examination types)? While clearly the risk would increase by around a factor of 3–4, the overall revised risk of approximately 90–150 cases per million women examined (compared with 30–50) is still arguably very low; however, to the women in that group of 40 to 100 additional women whose cancer would be induced at the higher level exposure, this is an important change, particularly if it indicates sloppiness and center-dependent variations when it comes to radiation control measures. An indicator of success with breast screening programs is high attendance rates among at-risk women—greater than 70%, for example—and if women learned that the control on radiation doses was not as rigorous as it could be, it could impact attendance rates.

Other reasons for optimizing radiation delivery in diagnostic imaging departments include:

1. Radiation, while generally poorly understood, does introduce (perhaps a disproportionate level of) fear. Therefore, if the radiologic community can demonstrate rigid controls on radiation levels, it should ultimate reduce patient anxiety *and* parent/caregiver anxiety.
2. The background data justifying the need for diagnostic reference levels already show that variations between departments for the same examination and similar patient size are excessive, and radiation protection standards must be rigorously applied across all departments to make sure that reported dose variations are minimized.
3. Some radiation doses are not insignificant, as shown elsewhere. Therefore, the stochastic and deterministic risks for the relatively high-dose examinations should be kept to a minimum wherever possible.
4. If we do not implement radiation protection standards, the most vulnerable of our society is at greatest risk. Children have a greater radio sensitivity than adults, and since they would normally live for a greater number of years post-exposure, they have more opportunity to express any induced cancers.
5. In many regions, including the European states, United States, Canada, and Australia, it is a legislative requirement that radiation protection standards should be implemented.

The ICRP suggest that the implementation of ***diagnostic reference levels*** is a key way of optimizing exposures in medical imaging, something with which the authors totally agree. However, over-reliance on reference levels is not good, since showing that one hospital's radiation dose for CT head is below the acceptable 75th percentile value may fail to demonstrate this dose results from an optimized procedure, as it may be possible to further reduce dose while not affecting the diagnostic efficacy of the examination (or conversely, the dose could be too low to produce a suitable image for an adequate diagnosis). The authors of this text are not convinced that the imaging community has been as diligent as it could be to ensure that doses to patients are as low as reasonably achievable. Three contexts will be given to support this argument, each focusing on one of three common reasons for non-optimization: (1) Reliance on traditionally

employed, well established procedures; (2) acceptance of new technologies; and (3) non-implementation of findings in the literature that might support lower doses.

Why Might Non-Optimization be Evident?

Reliance on Traditionally Employed, Well-Established Procedures

In radiography, we have been fortunate to have a number of important and highly influential pioneers who have led the way in the area of devising and publishing effective positioning and technical criteria for most examinations performed day-to-day. One such person was Kitty Clarke, (**Figure 5-5**) who not only founded one of the first schools in radiography in the 1930s, but published the seminal textbook *Positioning in Radiography* in 1939, which transformed the radiographic practice and is still being published in a revised form today.

While the importance of that textbook is not debated, fundamental components of techniques that were proposed in 1939 may not have the same relevance or need as today. Let us look at one example—the distance of the x-ray source to the image receptor (or film as it was in 1939), which for many non-erect positions was recommended to be 100 cm. In 1939, this distance was possibly more necessary than today due to the lower power available to x-ray tubes, and therefore the need to keep the x-ray source within close vicinity of the receptor, but with today's technology, is this distance necessary? And more important, will maintaining it result in an optimized examination? Let us consider the science for a moment.

Figure 5-5 Kitty Clarke, a radiographic pioneer.

Geometric un-sharpness is one of the key sources of poor definition within radiographic images. It arises from the finite size of the focal spot that can be anywhere between 0.1 and 2 mm, which leads the edge of an object being displayed as a slightly blurred entity (***penumbra***) as opposed to a very sharp point. This is displayed in **Figures 5-6** and **5-7**. The size of this penumbra is dependent on three key factors: (1) The size of the focal spot; (2) the distance between the focal spot and the object being irradiated; and (3) distance between the object and the image receptor.

This can be summarized as:

$$U_g = \frac{S_{fs} \times D_{o \to i}}{D_{S \to o}}$$

where U_g is geometric unsharpness, S_{fs} is focal spot size, $D_{o \to i}$ is distance from the object to the image receptor, and $D_{S \to o}$ is distance from the focal spot to the object.

The important thing to note here is that as the focal spot to object distance ($D_{S \to o}$ above), *increases*, the geometric unsharpness *decreases*; in other words, as we increase the distance between the x-ray source and the patient (and other factors remain unchanged), the image should get sharper. (Whether this will improve diagnostic efficacy in all situations would require examination specific observer studies.) One way of increasing the source to patient distance ($D_{S \to o}$) is by increasing the distance from the x-ray source to image receptor, since the patient to image receptor distance should remain constant.

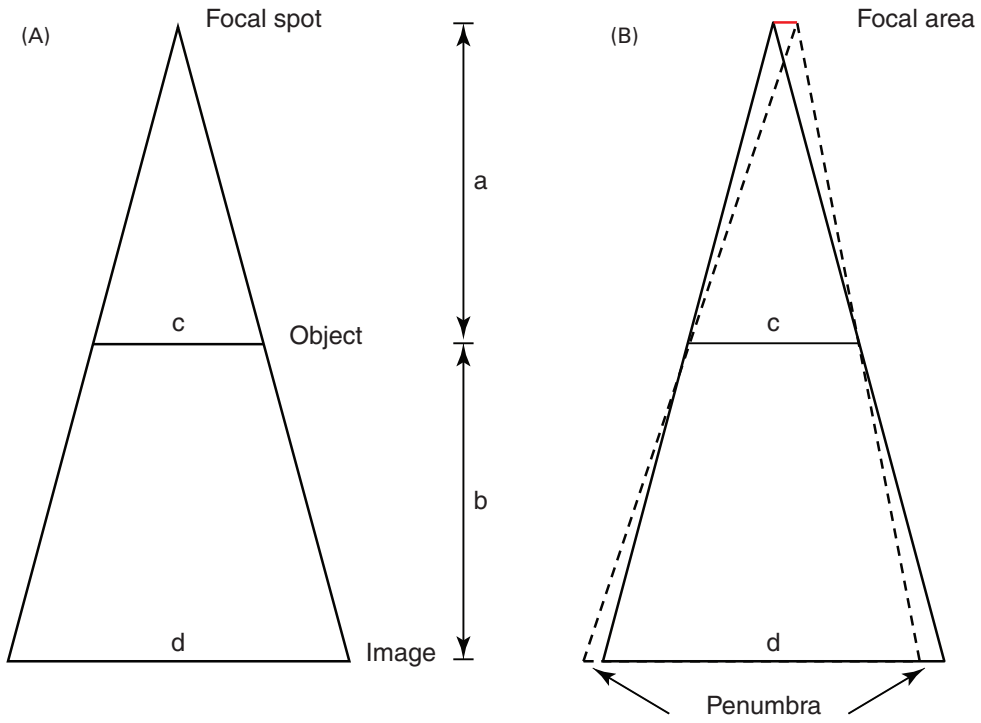


Figure 5-6 Geometric unsharpness. In the figure on the left (A), the focal spot size is infinitely small and therefore the reproduction of the edge of the object being irradiated is very sharp. On the right (B), the focal spot area demonstrates a definite width resulting in a penumbra or blurred reproduction of the edge.

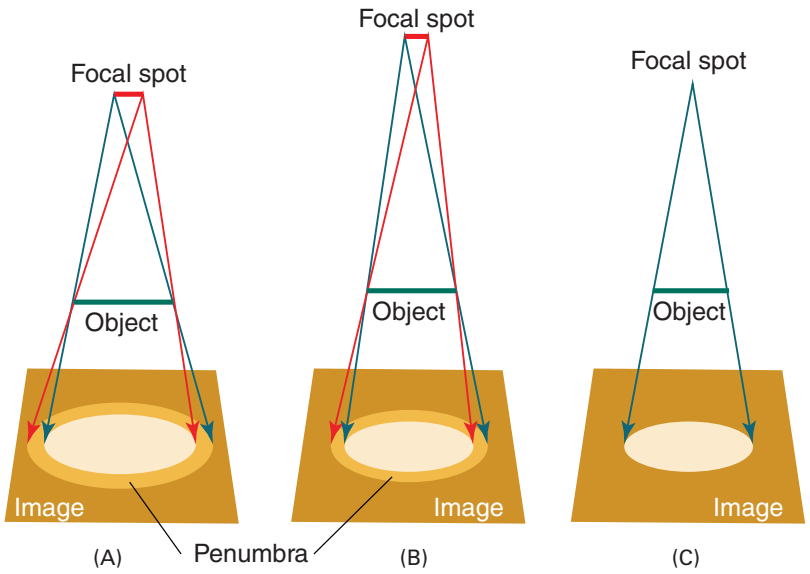


Figure 5-7 Illustration of the impact of increasing the source-to-image receptor distance on image quality. In A, the penumbra can easily be seen resulting from the geometric unsharpness generated by the focal spot size. In B, this unsharpness has been reduced as a result of the increased source-to-receptor distance; however, a penumbra is still evident when compared with an ideal infinitely small focal spot size (C).

While the positive implication for image quality is very evident when one uses a longer source-to-patient distance (by increasing the source-to-receptor distance), it can be asked what this has to do with optimization, since there has been no reduction in dose. There are two answers to this: first, if we go back to the European definition of optimization a few paragraphs above, we see that it does not automatically say that *reductions* in doses will be achieved; rather, that they will be kept *as low as reasonably achievable* while obtaining the required diagnostic information. The second half of this means that *providing the information for an accurate diagnosis is fundamental to optimization*, and techniques that involve increasing the source-to-patient distance, potentially improving image quality and better defining pathologic appearances, should be investigated rigorously. Second, as it happens, a dose reduction advantage with this technique *has* been argued for in the literature due the effect of the Inverse Square Law, where an effective dose reduction of 33% has been observed when the source-to-receptor distance is increased from 100 cm to 130 cm (while the patient-to-receptor distance remained constant) for pelvis examinations. While the extent of the dose reduction associated with increasing the source-to-patient distance is debated (Huda et al. 2005; Brennan and O'Leary 2006), and an awareness of the type of collimation that needs to be used to reap the full benefits of this technique is required (Poletti and McLean 2005; O'Leary and Brennan 2006), the potential overall benefit to the patient of increasing the distance (which incurs little or no additional cost) should have led to a rigorous reevaluation of the traditional distance of 100 cm. This has not happened.

It should be acknowledged that with an increased source-to-receptor distance, one needs to consider the constraints imposed by table unit position, grid focus specifications, and the height of the radiologic technologist!

Acceptance of New Technologies

When new technologies are being proposed, it is important that they are introduced in a careful way so they are used to their full potential at the lowest risk to the patient. There may be a flurry of research activity, often supported by manufacturers, to demonstrate the efficacy of new equipment, but this does not usually translate into regular departmental or even national reviews of the technology to ensure that optimal usage is in place, and that the technology is indeed useful to the specific environment in which the equipment is placed.

Two examples might help—the first around the issue of non-optimized use. The introduction of digital technology is probably the greatest change within medical imaging that we have seen since x-rays were discovered in 1895. A lot of important research was initially performed—for example, the DMIST trial in mammography (Pisano et al. 2008)—to make sure that the new technology offered benefits to the patient and clinicians. However, while the efficacy of digital acquisition was made clear, regular follow up studies are rarely evident to make sure that computed (CR) (**Figure 5-8**) or direct radiography

Figure 5-8 Computed radiography system.
Courtesy of Fuji

Figure 5-9 Direct radiography system. X-ray tube and table (A). DR battery and charger (B). DR cassette (C).

(DR) (**Figure 5-9**) systems with a variety of *potential* benefits are being used in a way that maximizes (or even realizes) those benefits.

Specifically around exposure factors, the reader may be aware that the materials used in digital receptors and their attenuation properties are quite different from those contained within rare-earth intensifying screens and films. The materials used in rare-earth screens had K-edge values of around 52 and 39 for gadolinium- and lanthanum-based phosphors, respectively, compared with values closer to 37, 35, and 13, respectively, for barium-, cesium-, and selenium-based materials used with digital acquisitions in CR (barium) and DR (cesium and selenium). If we wish to maximize the attenuation of x-rays by the image receptors, we should be using x-ray beams with energy profiles so that the mean, median or effective beam energy emerging from the patient is at or just above the K-edge of the receptor materials. In other words, due to the generally lower K-edge with digital technologies compared with intensifying screens, it is possible that the kVp selection should be lower than that used with screens. Of course, the potentially higher skin doses with lower energies would need to be investigated, but there appears to be a paucity of clinical-based research investigating this, and therefore exposure factors traditionally used for many years with film are often being used in the context of digital imaging. If we are matching emergent energies with K-edge of the receptors in a more efficient way, this would mean greater attenuation of x-rays and potentially dose savings, since the x-rays would be captured more effectively. It is this level of complexity that often exists within imaging that demands ongoing, clinically relevant research. Until this work is done, we cannot be sure that the exposures being employed are optimized and that the potential dose savings or image quality enhancements often quoted with digital technology are being maximized.

The second example addresses the issue of making sure that the new technology is suitable to each environment in which it is placed. Initial efficacy research cannot examine all possible application situations for novel equipment. When digital breast tomosynthesis (DBT) was being introduced, there was a plethora of research published demonstrating DBT's benefits in terms of sensitivity and specificity, with some of the most impressive and important work being performed in collaboration

with industry (Skaane et al. 2013). At the time of this writing (2013), due to these research outputs, in the opinion of the authors, it is likely that DBT (quite rightly) will play a critical role in breast imaging in the future and will have an important place within breast screening strategies. However, without debating the potential value of DBT, specific benefits to women in particular environments are as yet unclear. For example (again at the time of writing), it appears that DBT is being introduced in the Middle East, a region in which the age of the women being screened and the nature of their breast tissue, as well as the profile of the cancers encountered in this population, are quite different from these factors in women on whom most of the DBT studies have been performed. Research needs to be done on the specific populations who will be utilizing this new technology to see if the benefits described elsewhere are relevant to these different groups of women. In addition, the technology may be introduced into new environments within developing countries where radiologists and radiographers may not have the same level of expertise or training as those involved in the original research studies; ongoing reviews and work are necessary to ensure that women being examined or not being affected in a deleterious way following the technology change. Simple acceptance of new technology without the relevant, tailored, supporting evidence should not happen.

Finally, research-funding agencies must take some responsibility for these issues of optimal usage of equipment and suitability of new technologies for different populations. In most environments there is (quite rightly) much emphasis on providing monies to support studies focusing on innovative ideas; unfortunately, the same emphasis is not placed on making sure we are getting the most benefit out of the new technology at the lowest patient risk once this new technology has been installed. Until this imbalance is addressed, it is difficult to see how one can make sure of maximum diagnostic yield at lowest radiation risk to the patient.

Non-Implementation of Findings in the Literature

Again, a simple radiographic example may be useful here. A study performed some time ago demonstrated the potential benefit of performing a PA rather than an AP projection for the abdomen (Brennan and Madigan 2000). While this proposed change in technique could clearly only be performed on reasonably mobile patients, the benefits were significant, with reductions of around 40% in patient entrance surface dose and internal phantom dose, respectively, and no change in image quality. The reason proposed for this reduction in dose was the decreased patient diameter due to tissue displacement and the radioprotective nature of anatomy located at the posterior part of the patient. While this technique had no apparent cost or social implications, widespread adoption of this technique has not occurred.

Poor translation of research findings into practice within medicine is not uncommon. It appears that performing scientifically valid studies and publishing resultant findings in important peer-reviewed journals is not enough; systems or process must be in place to make sure that the latest important findings relevant to a particular discipline are regularly reviewed and implemented whenever appropriate. The value of bodies such as NICE has been considered earlier in this chapter, and this type of arrangement—albeit on a smaller scale—needs to be in place within radiology and radiography to maximize translation of best practice. This body would have the potential to take responsible for reviewing and implementing current and new technology and practice and should consist of experts from a variety of bodies including consumer groups facilitating a multidisciplinary. Sadly, at this time, such organizations are not common.

Optimization Initiatives

Some national initiatives that are focused on image optimization are in place in different countries and are responsible for highly effective initiatives such as the implementation of diagnostic reference

levels in the UK by the Health Protection Agency, formally known as the National Radiological Protection Board. Other bodies include Image Gently in the United States, which focuses on optimizing procedures for children; NEXT in the United States, which looks at keeping radiation risks as low as possible for a variety of examinations; and ARPANSA, which looks at CT reference levels in Australia. These are the type of activities that will lead to important optimization benefits within radiology and radiography, but should be evident more widely if widespread optimization is to be in place.

Optimization and Cost

Before implementation of any new optimized procedure or equipment—even if significant dose reductions are shown and/or improved image quality—two issues that are referred to in the European legislation should be considered: *economic and social factors*. Economic clearly means how much it would cost; while the increased source-to-receptor distance example shows that in some cases, the cost is minimal (apart perhaps from some increased wear and tear on the x-ray tube due to having to increase the exposure at the greater distances), the cost for other techniques may be more significant. An interesting example of how a cost-benefit analysis can be performed across a country and what parameters should be included in such an analysis was provided by Ginsberg et al. (1998) when the use of rare-earth screens was being proposed for widespread use in Israel. While the context is outdated, the approach is still very relevant today. The social factors are harder to define, but the acceptability of introducing the technique from a clinician and patient perspective must form an important part. To return to our example of the increased source-to-receptor distance: in implementing this change, the new technique should present little or no difficulty for the patient; however, for staff, there is the challenge of smaller staff members being able to raise the x-ray tube to a level to accommodate the increased source-to-receptor distance. While this specific challenge should be easily overcome with the availability of a step or similar tools, other optimizing techniques may present much more challenging circumstances. All costs and social considerations must be carefully examined before clinical implementation.

Optimization: Image Quality Versus Diagnostic Efficacy

Finally, for clarity, the author would like to differentiate between ***image quality*** and ***diagnostic efficacy***, as both these terms have been used in this chapter. Increased image quality refers to a circumstance in which the appearance of the image shows an improvement, either subjectively to the observer or measurably, which could be related to increased contrast or sharpness, for example. However, increased diagnostic efficacy goes a step further: It describes a situation where there is now an *increase in the level of clinical information that is provided* to the radiologist or other expert observer so that diagnosis is improved. While increased image quality improves the likelihood that diagnostic efficacy will be improved, it is important to be clear that for a variety of reasons, this is not always the case.

Radiation Dose Limits

Obviously, imposing dose limits is another way to limit radiation risks; however, this is not an option for patients undergoing medical imaging. The bottom line is that if a diagnosis is needed for the future well being of the patient, the radiation that is required must be administered. In fact, a dose limit could potentially be an obstacle to diagnosis.

On the other hand, there are dose limits for workers and members of the public (perhaps family members accompanying patients) that should be adhered to. Those published by the ICRP and other national or international agencies are summarized in **Table 5-5**. If these values are exceeded, appropriate measures should be put into place to minimize

Table 5-5 Dose limits for workers and members of the public

Measure	Occupational exposure	Members of the public
ICRP (2007)		
Effective dose	20 mSv per year, averaged over 5 years	1mSv
Lens of the eye	150 mSv	15 mSv
Skin	500 mSv	50 mSv
Hands & feet	500 mSv	—
United States (REF AND DETAILS, NEED TO BE ADDED)		
Effective dose	50 mSv	1mSv
Lens of the eye	150 mSv	15 mSv
Skin	500 mSv	50 mSv
Hands & feet	500 mSv	50 mSv
Under 18 year olds		
Effective dose	1 mSv	—
Lens of the eye	15 mSv	—
Skin	50 mSv	—
Hand & feet	50 mSv	—
European Union (1996)		
**Workers, apprentices and students (aged 18 or over)*		
Effective dose	100 mSv over a five year period, maximum of 50 mSv in any one year	1mSv
Lens of the eye	150 mSv	15 mSv
Skin	500 mSv	50 mSv
Hands, forearm, feet & ankle	500 mSv	—
Apprentices and students (aged 16 & 17)		
Effective dose	6 mSv	—
Lens of the eye	50 mSv	—
Skin	150 mSv	—
Hand & feet	150 mSv	—
Australia (2008)		
Effective dose	20 mSv per year, averaged over 5 years	1mSv
Lens of the eye	150 mSv	15 mSv
Skin	500 mSv	50 mSv
Hands & feet	500 mSv	—
Canada (2008)		
Whole body (Effective dose)	20 mSv per year, averaged over 5 years	1mSv
Lens of the eye	150 mSv	15 mSv
Skin	500 mSv	50 mSv
Hands	500 mSv	50 mSv
All other organs	500 mSv	50 mSv

Key: * = In the European Union, workers are classified into Category A or B. Category A workers are those who could receive an annual effective dose greater than 6 mSv or an equivalent dose to the lens, skin or extremities that is larger than 3/10s of the limits described within the table. Category B are all other workers

Unless otherwise stated these are annual values. Apart from the effective doses, other values are dose equivalents.

reoccurrence, and the exposed individual should be monitored for any adverse sequelae.

Obviously, special consideration must be given to pregnant patients requiring medical diagnostic exposures. It is important to note that there is no evidence that exposure to an unborn child from diagnostic radiologic examination presents any risk to the child of pre- or postnatal death, growth malformations, or mental impairment (ICRP 2007). In fact, the risk to an unborn child of a stochastic event is the same as any young child (ICRP 2007). Nonetheless, to minimize the risk, it is important that a patient informs the radiologist or radiographer prior to any exposure so that special consideration is given to optimize the procedure. A potential procedure would be as follows: If an x-ray procedure is likely to provide a radiation dose of more than, for example, 1 mSv to the unborn child, then specific justification for the need of that examination needs to be performed along with an assessment of (1) the risk to the unborn child if the examination is performed and (2) risk to the woman if the examination is not performed. If it is decided that the examination should go ahead and an optimized procedure is identified, the risks to the child are usually explained to the women and to the referrer of the examination *before* the examination has been performed. The estimated radiation dose delivered will be recorded.

Radiation Detection and Measurement

The principles of radiation protection—justification, optimization, and dose limits (particularly the last two)—rely on effective methods on measuring radiation dose. There are a number of ways of doing this, but the main methods used in diagnostic imaging departments are:

- thermoluminescent dosimetry;
- dose-area product technology
- solid-state meters
- radiochromic film

Each of these will be considered in turn. It is important to emphasize that the dose quantity measured with most current commercial units used in diagnostic imaging departments is the air kerma.

Thermoluminescent Dosimetry

Overview

Thermoluminescent dose (TLD) meters can be used for patient and staff radiation measurements. They are made of a variety of materials but commonly come as lithium fluoride or lithium bromide, and in recent years, some have been doped with copper or manganese to increase the sensitivity to radiation. They are usually small chips around a few millimeters in diameter, but are available in other sizes and in powder form (**Figure 5-10**).

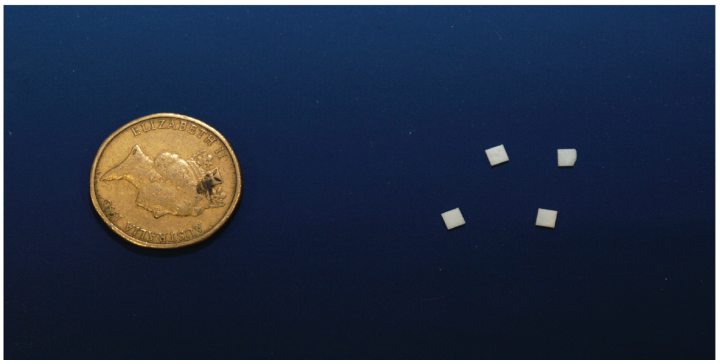


Figure 5-10 Thermoluminescent dosimeters beside an Australian dollar coin. The coin is about 2 cm in diameter.

This range of sizes offers much versatility and facilitates measurements on the entrance surface of patients, for example, without any adverse image quality effects, as well as at a range of body locations for staff such as the tips of fingers. They are also useful for measurements at a variety of locations inside an anthropomorphic phantom.

TLDs: How Do They Work?

TLDs work on the principle that when they are exposed to ionizing radiation such as x-rays or gamma rays, electrons within the dosimeter crystals are moved to a higher energy level and are trapped there as a result of deliberately added impurities such as manganese. The electrons remain in this higher-energy state until heat is applied, whereupon they move back to their normal position and release the excess energy in the form of light. The number of light photons that are released are counted, and following appropriate calibration procedures, the radiation dose can be calculated based on light photon number. TLDs offer the advantage of being able to measure backscattered radiation from the exposed individual, which may account for up to an additional 40% of dose (air kerma).

Modern TLDs such as lithium fluoride doped with manganese (LiF: Mg, Ti) have the capability to record doses as low as 10 μGy and as high as 1 Gy, whereas copper-doped versions (LiF; Mg, Cu, P) can extend this range from 1 μGy to 20 Gy. TLDs can record x-ray photons with energies above 5 keV.

Patient Measurements

For patient dose measurements, TLDs are usually placed on the patient surface at the central entrance point of the radiation (**Figure 5-11**).

In case of erroneous recordings of dose, sometimes two or three TLDs are positioned at the same time. They are placed in a black plastic sachet, which protects the TLD from dirt and grease, but more importantly represents the stratum corneum of the skin, which is made up of dead cells. This means that the radiation amount that reaches the TLD represents the radiation amount that would have reached the first living layer of patient tissue. From this position, the entrance surface dose to the patient can be calculated, from which effective doses can be determined.

Figure 5-11 TLD placed on the entrance surface of the patient at the center of the x-ray beam.

Patient measurements with TLDs can only be used effectively for non-fluoroscopic examinations. During fluoroscopic procedures such as barium swallows or cardiac angiography, the patient moves considerably throughout the examination, and since the TLD is usually on a fixed patient position, this means that the TLD will change its distance from the x-ray source—it may sometimes be on the entrance surface and other times on the exit surface, and may even be outside the x-ray beam's field for some of the exposure. This allows for much misinterpretation when faced with resultant dose values; a better alternative is the dose-area product (DAP) meter discussed below.

One other examination type that TLDs are not suited for is mammography. In mammography, the beam energies are traditionally much lower than in other fields of radiography, meaning that there is the possibility that the TLD will be seen on resultant images. This would present with certain challenges, since breast pathologies such as microcalcifications are often very subtle, and the superimposition of a TLD image may have serious deleterious effects.

Staff Measurements

For measuring staff doses, the TLD is placed in a badge-like structure (**Figure 5-12**), which can come in a variety of forms.

One type contains two TLD discs, one thin (40 microns) and one thick (90 microns). The thin disc has no plastic badge covering and allows the measurement of low-energy doses, while the thicker disc is covered by a thick layer of plastic, which facilitates the recording of higher energy exposures. Under normal circumstances, the badge is worn close to the gonadal areas, and when lead rubber aprons are being worn, the dosimeters are worn under the protective apron to better represent the radiation dose actually being received by the organs. The doses received are usually very low, and in the large majority of cases, there is no reading on the TLD. During interventional procedures, where doses are generally higher and exposure times are longer compared with other fluoroscopic examinations, staff may wear a further TLD dosimeter close to the thyroid if a protective collar is not worn; indeed, the Health Protection Agency in the UK has a dosimeter that can be positioned at the thyroid that is calibrated to assess radiation dose levels to the lens of the eye. It has been proposed that TLDs positioned in regions of the body that are not usually covered by protective garments are possibly more valuable that those positioned underneath aprons, where the dose is nearly always at an immeasurably low level. A recent survey of radiology departments in 13 European countries demonstrated that the practice in five of these countries was to wear a single TLD outside the lead apron at the position of the collar.

Figure 5-12 Thermoluminescent dosimeter badge.

Calibration

Each dosimeter must be calibrated to facilitate useful TLD dose measurements; otherwise, there is no way of knowing how many light photons equals how much radiation dose. To do this, before any patient or staff exposure takes place, the TLDs are irradiated using a known level of radiation dose, usually a batch at a time. The light photons released are then counted, and a ratio between light photons and the known radiation dose is calculated, which then facilitates all future calculations. After this procedure, the TLDs are cleared of any remnant excess energy before they are used to measure patient or staff doses. It is critically important to note that TLDs are calibrated at a specific x-ray beam, which means that they should only be used at or close to this specific energy.

TLD Reading

Following exposure, the TLDs are then heated (*thermo*luminescent) to release light (thermo*luminescent*). This is done using a TLD reading oven (**Figure 5-13**) using hot nitrogen gas.

Figure 5-13 Thermoluminescent dosimeter reader.

Traditionally, TLD reading was a very tedious process, with only one TLD being read at any one time; however, modern units can now read up to 280 dosimeters within an hour. Within each heating oven, there is a particular cycle that consists of a pre-heating stage (up to 165°C), a reading stage (up to 300°C), and an annealing stage. The aim of the latter stage is to clear the TLD of any remnant recording of dose so that when used for a specific patient measurement, the vast majority of the reading on the TLD results from that patient exposure.

Uncertainty with TLD Measurements

While TLDs for patient measurements are cheap, around $5 each, anyone who has used them will be aware that there are a number of uncertainties around their measurements. These uncertainties arise from a variety of sources and include:

1. **TLD signal fade**. This is where dose information that is stored on the TLD in the form of electrons trapped at higher energy levels starts to decrease even before reading of the TLD. Depending on the type of TLD, the type of exposure and the type of annealing process, the level of fading can vary between 1% in a year to 7% in the first couple of weeks.
2. **Non-linear response of TLDs to radiation dose**. This is particularly a problem at the lower and higher dose readings, but manufacturers would argue that with modern-day TLDs and readers, the linearity is within 5% for doses between 10 μGy and 1 Gy.
3. **Dependency of TLDs on radiation beam energy**. TLDs are highly sensitive to the level of energy to which they are exposed. This means, for example, that if a batch of TLDs is calibrated at 80 kVp, then strictly speaking, one should only use the TLDs close to this beam energy. In practice, however, this adherence to a single energy value is difficult, since patients come in a variety of sizes and conditions and therefore require a variety of energy settings.
4. **The validity of the calibration**. As mentioned above, TLDs must be calibrated before use, so that the radiation dose that the patient (or staff member) has been exposed to can be calculated from the number of light photons being released. However, this is far from a perfect science; for example, when it was said above that for calibration TLDs are exposed to an *known level of radiation dose*, that known dose is only as reliable as the measuring provided by a unit such as a solid-state dosimeter. These dosimeters are prone to error and need to be regularly calibrated to some primary or secondary source to make sure that their dose readings are valid.
5. **Variations in TLD reader performance**. Unfortunately, if two TLDs are exposed to the same level of radiation, one will not necessarily get the same reading from each dosimeter if they are read by different machines. The performance of the reader can vary depending on age, the rigor and recency of calibration, the model type and level of precision offered, and the technical support.

These sources of error are not insignificant, but for radiation doses between 0.1 mGy and 11 mGy, overall uncertainty should not be larger than 25%

at the 95% confidence level between beam energies of 50 kVp with 2.5 mm Al equivalent and 120 kVp with 5 mm Al equivalent. On the one hand, it is reassuring that these conditions should cover *most* exposure conditions encountered within diagnostic imaging centers; on the other hand, an uncertainty of 25% means that real changes or differences in radiation doses between patients or experimental conditions would need to be sizable if they are not to be obscured by uncertainty. Also, the lower dose at which any reasonable level of certainty remains (0.1 mGy) is actually not that low, and certainly this dose and below would be relevant for some chest and pediatric imaging and would be close to values demonstrated for exit patient doses and internal phantoms measurements. An often-quoted solution to this low-dose inaccuracy (or even sometimes non-reading) is to expose the same TLD to several exposures across 5 or 10 patients and then take the mean value; however, this averaging approach comes with its own difficulties.

Some of these uncertainties may be alleviated by a new technology: optically stimulated luminescent dosimetry (OSL). These are made of aluminum oxide crystal detectors (Al_2O_3) that, following radiation exposure, can release details on the dose received by using *light* stimulation and not heat (**Figure 5-14**).

Figure 5-14 Optically stimulated luminescent dosimeter (OSL).

Following irradiation, the dosimeter is stimulated by a laser (green) light, and the emitted blue light is then amplified using a photomultiplier tube and send on for absorbed dose estimation. While a number of the inaccuracies and inconveniences associated with heating the TLDs are removed, the other key advantage of OSLs are that they can be placed on the site of investigation in mammography without any effect on image display. The use of OSLs can be seen in **Figure 5-15**. The apparatus in this case being designed to facilitate real-time dosimetry readings in mammography (Aznar et al. 2005): the OSL is attached to the readout and display electronics using fiber links and is stimulated using a green laser light.

Dose-Area Product Meters (DAPs)

Overview

Dose-area product dosimetry is possibly the most common method of patient dosimetry currently being carried out. The arrangement requires an ionization chamber to be attached to the output of the x-ray tube (**Figure 5-16, A** and **B**); however, the facilitation of readings that are generated automatically following exposure (**Figure 5-17**) makes this a highly attractive option, where placement of dosimeters on patients and elaborated reading processes are not required.

Indeed, in the European Union, it is a requirement that this type of dosimeter should be available with all new x-ray equipment:

> *If new radiodiagnostic equipment is used, it shall have, where practicable, a device informing the practitioner of the quantity of radiation produced by the equipment during the radiological procedure. Euratom 97 43 (1997).*

DAP Meters: How Do They Work?

DAP meters, as the name suggests, simply relies on measuring the amount of radiation interacting with the ionization chamber (located at the x-ray tube) (**Figure 5-16**) and multiplying this value by the area of exposure. Resultant doses are expressed

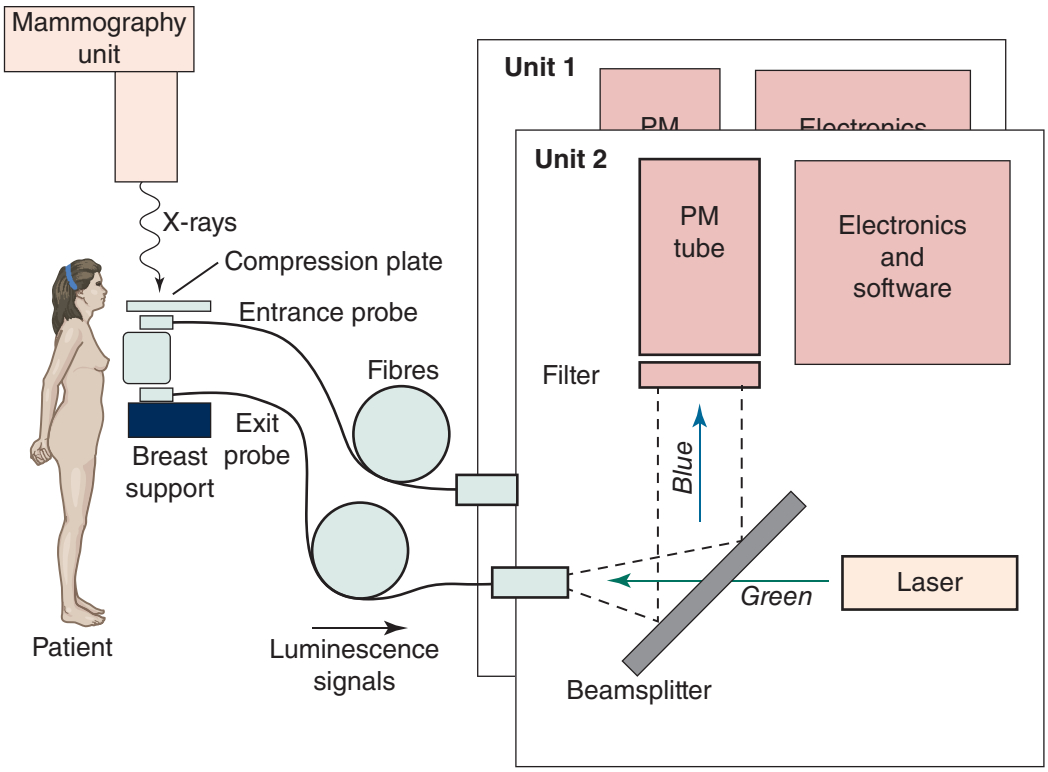
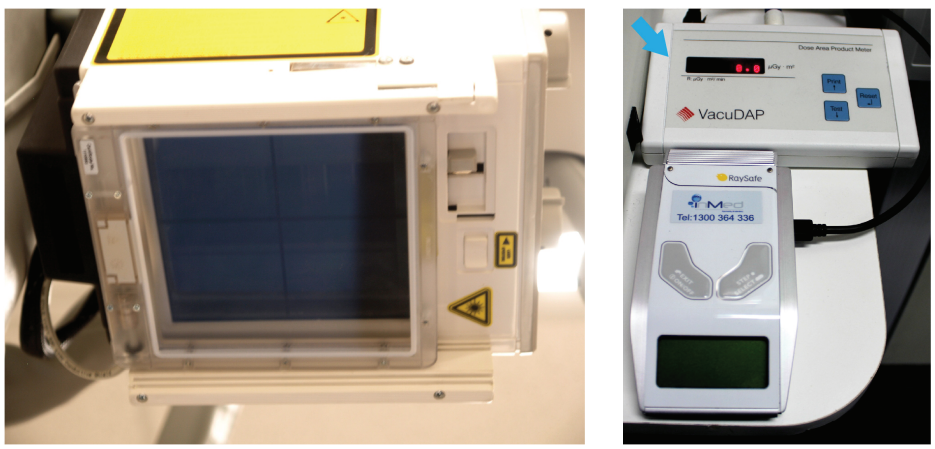


Figure 5-15 An example of the placement of OSLs in mammography for real-time dosimetry. An OSL is placed at the entrance and exit surface of the breast, and these are attached to the read and display electronic using fibers.

Reproduced from: http://bjr.birjournals.org/content/78/928/328.figures-only; Aznar MC, Hemdal B, Medin J, Marckmann CJ, Andersen CE, Bøtter-Jensen L, Andersson I, Mattsson S. In vivo absorbed dose measurements in mammography using a new real-time luminescence technique. Br J Radiol. 2005 78(928):328–34.



(A) (B)

Figure 5-16 Photograph of a DAP ionization chamber (A) and readout unit (B, see arrow).

Figure 5-17 DAP readout unit (see arrow).

using the unit of Gycm^2; however, modern units in addition simultaneously record DAP rate and irradiation time. The ionization chamber is simply a radiation detector, which contains air, the particles of which require energies of around 34 eV to ionize, thus creating an ion pair (**Figure 5-18**).

So, for example, if the chamber is exposed to a photon with an energy of about 34 keV, approximately 1000 ion pairs will be created—1000 electrons and 1000 positive ions—and these will be attracted to the anode and cathode, respectively. In reality, the electron output from the ionization chamber is very low, and this signal is therefore amplified before the dose data is sent to the display device, which informs the operator.

Location of the Ionization Chamber and Associated Advantages

Since the ionization chamber is attached to the output of the x-ray tube, it must be as transparent as possible so that it does not attenuate too many x-rays, nor interfere with the light beam diaphragm device. However, it is this attachment to the x-ray tube that makes it so versatile and follows the x-ray tube wherever it is placed. In addition, the location of the ionization chamber at the x-ray tube, but at the patient side of the collimators facilitates two important requirements: firstly that the ionization chamber is positioned perpendicular to the x-ray beam; secondly that the ionization chamber captures the whole area of the x-ray beam, in other words however large the collimated field, the area of the ionization chamber is always larger than this. As mentioned above, the resultant dose value given is therefore the product of the absorbed dose at the chamber and the area of exposure, so for example if a dose of 5 mGy is delivered over an area of 2 cm × 2 cm, the dose-area product (DAP) reading will be 20 mGycm^2. If, however, the area is increased to 4 cm × 4 cm

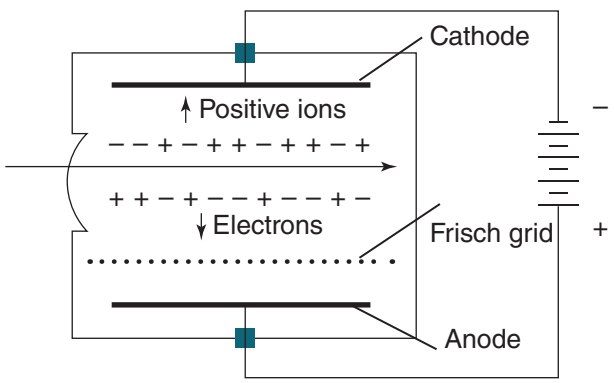
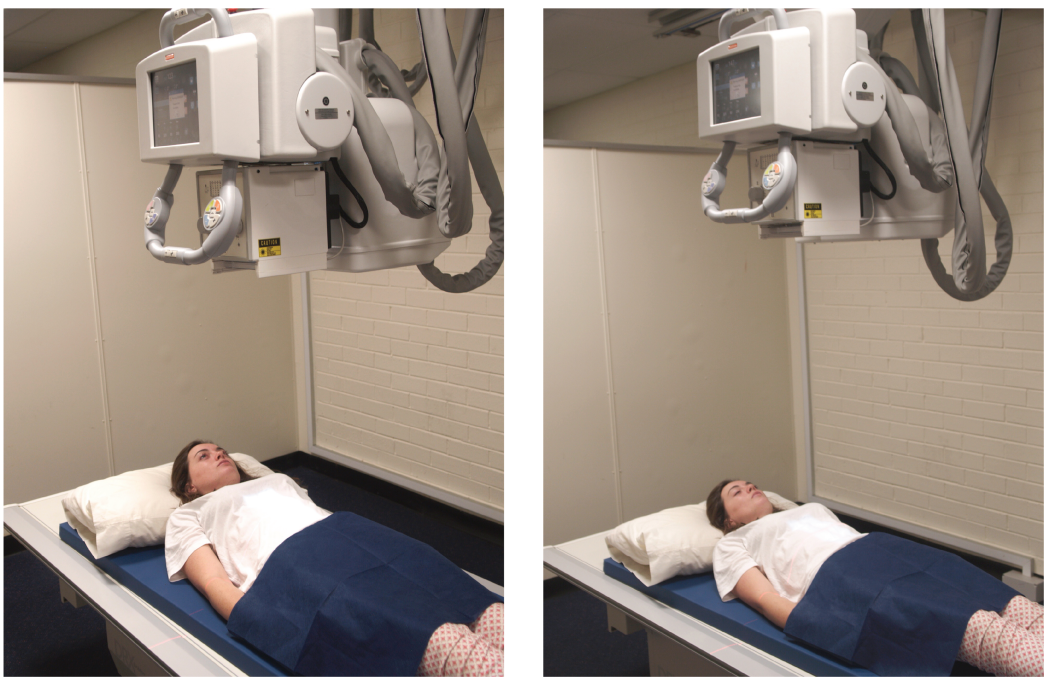


Figure 5-18 A diagram of a part of an ionization chamber. Some ion pairs are shown and these will be attracted to one of the two electrodes.

and the dose remains the same, the DAP reading is now 80 mGycm^2. One key consequence of measuring the DAP in this way and at the x-ray source is that, regardless of the distance that the patient is from the x-ray tube, the radiation dose will remain constant if the beam is appropriately collimated.

Here is an example:

Imagine a patient at 100 cm from the DAP's ionization chamber and the area of exposure *at the DAP meter* required to expose the abdomen is 4 cm × 4 cm. The absorbed dose at the DAP is 2mGy, resulting in a DAP reading of 32 mGycm^2. If the distance between the patient and the chamber is increased to 200 cm, because of the nature of the divergent beam, the collimation field will have to be made smaller to ensure a constant x-ray field on the patient's surface, otherwise the whole patient could be irradiated. In fact, the field would have to be reduced to 2 cm × 2 cm. However, since the patient's distance from the x-ray tube has doubled, the dose at the patient has been reduced by a factor of 4 in accordance with the Inverse Square Law; therefore, to maintain an adequate dose at the patient, the exposure and hence the dose at the chamber will have to be increased by a factor of 4, to 8 mGy. At this new patient position, simple calculations will demonstrate that the DAP reading will be 32 mGycm^2, which is identical to the original exposure at 100 cm and reflects the fact that because of the interdependence of radiation dose and area of exposure as distance changes, the dose at the entrance of the patient has remained the same, regardless of the distance. A simple dose reading at the chamber without considering the area of exposure would not have been adjusted in the same way. A photograph of a patient being examined at two distances is shown in **Figure 5-19**.



(A) (B)

Figure 5-19 Increasing distance for an AP lumbar spine examination. (A) normal distance (100 cm); (B) increased distance (130 cm).

Traditionally, the connection between the DAP's ionization chamber and the display device that demonstrates the subsequent readings had to be long enough to facilitate positioning of the x-ray tube in all possible locations within the x-ray room; however, modern versions using wireless Bluetooth technology have provided a technological solution to this challenge.

DAP Limitations

Because readings happen with minimum input from the radiographer or radiologist (making them very convenient) and the inclusion of the area of exposure as well as the dose, it is likely that this form of dosimetry will be the main method used for widespread patient dosimetry of the future. However, DAPs do not come without their limitations. First, DAP meters need to be regularly calibrated. A study performed in UK hospitals (Crawley et al. 2001) demonstrated that the DAP reading given on 31% of the 41 units measured was more than 10% away from the true DAP value, with miscalibration being more apparent for under-couch (50%) rather than over-couch (23%) tube locations. There were no differences between the level of miscalibration between the DAP units that came with the x-ray equipment and those fitted retrospectively. While it was acknowledged in the paper that air pressure and temperature could have, in theory, an impact of up to 5% on the readings, the units measured were highly unlikely to have been subjected to such extreme climatic variations. It was recommended that DAP calibrations should be performed at intervals of no less than 6 months.

The second limitation is around the actual measurement that is performed with DAP meters. Because of the location of the ionization chamber, the dose measured is that at the output of the x-ray tube and not that at the patient, and one must question how accurately the dose displayed therefore represents the actual patient dose. An example of where this can become a problem is when an object is placed between the DAP chamber and the patient, resulting in a DAP reading that implies a higher dose to the patient than what the patient actually received. This becomes an issue when under-couch tubes are employed and the x-ray table is positioned between the patient and the x-ray source, and attempts to measure patient skin or effective dose therefore require complex calculations.

A third limitation is that while DAP meters are useful for the calculation of the stochastic risks of the radiation exposures,, in recent years with increasing interventional doses, more emphasis is on skin dose. DAP readings have certain limitations here, since they do not distinguish between a large-field, low-dose exposure and a small-field, high-dose exposure. In other words, without further measuring devices such as TLDs or radiochromic film (discussed below), the exposure to specific skin sites is not properly assessed.

A final issue with the position of the DAP chamber is that subsequent readings cannot include the backscatter proportion (up to an extra 40%) that can be accounted for with TLDs placed on the patient's surface.

Nonetheless, while these limitations around DAP values representing patient exposures are acknowledged, work looking at the level of agreement between DAP readings and patient entrance dose when calculating effective dose have shown good agreement (Theocharopoulos et al. 2002). It is interesting to note also the conclusions of another study that compared the effective dose resulting from DAP and TLD measurements (Yakoumakis et al. 2001). Both methods were argued to be useful ways of calculating effective dose, although the DAP readings resulted in values that were up to 38% higher than those generated from TLD readings. The authors concluded that since the increased levels with the DAP readings were most likely related to the large exposure areas sometimes used by radiographers, resulting in fields that were in fact larger than the image detector area, DAP

readings may be a more accurate way of calculating effective dose.

Uncertainty with DAP Measurements

DAP measurements, like TLDs, are subject to certain uncertainties; however, it has been stated that these uncertainties should not be greater than ± 25% at the 95% confidence level for doses between 10 Gycm^2 to 10^3 Gy/cm^2 for x-ray energies between 50 kVp (2.5 mm Al) and 120 kVp (5 mm Al).

Solid-State Meters

Solid-state meters have the convenience of being placed wherever in the x-ray room one requires measurements along with the immediate display of dose data. They come as an electronic base unit, sometimes with additional probes; however, their size imposes certain limitations (the Unfors XI model pictured later in this chapter is approximately 14 cm × 7.5 cm × 3 cm).

Solid-state dosimeters have been around for a century. The modern types rely on semiconductor technology, the physics of which are beyond the scope of this text, but will be summarized here.

First, one must understand the terms ***valency*** and ***conduction bands*** within an atom. The outer orbital electrons in an atom are arranged in the form of an electron cloud. Some of these are tightly bound to the atom, and these are said to be located in the valency band, while others are free to move over reasonably large distances and are known to be located within the conduction band. The number of electrons contained within the conduction band will determine the object's conductivity; typically, metals such as copper have high numbers of electrons within this band. Insulators or semiconductors have no or very few electrons in the conduction band and therefore have low conductivity. The energy difference between electrons within these two bands is low, ranging from several eV to close to 1 eV in semiconductors.

Due to the low energy differences between bands, semiconductors may find some electrons within the conduction band, but the number of these electrons is amplified by the deliberate introduction of impurities within the structure, which serve as electron donors or electron receivers just below the conduction band. If a donor is provided, it means that electrons can move into the conduction zone at very low energies, and since the presence of electrons in the conduction zone facilitate conduction, this is known as a n-type (n = negative) semiconductor. Alternatively, the presence of a receiver encourages the movement of electrons from the valence band to this acceptor, leaving electron vacancies in the valence band which are known as holes; this is known as a p-type (p = positive) semiconductor. With both n- and p-type semiconductors the energy to create this electron movement is very small, ~1–3 eV, so irradiation of these devices can easily be detected at energy levels much less than those relevant to ionization chambers (34 keV). In practice, these n- and p-type semiconductors in circuits are joined together as shown in **Figure 5-20,** and any ionizations resulting from radiation interactions will result in an electrical current being created, which will indicate the level of the original irradiation.

Without getting into the intricacies of the potential semiconductor arrangements, one can see that semiconductors can serve as sensitive (and robust) detectors of radiation and have been proven to be highly efficient, with a response that is linear to the level of exposure. These types of dosimeters can be small in size, serving as a tool for personal dosimetry (**Figure 5-21**), or larger, making them relevant to much wider applications. (**Figure 5-22**).

Modern-type solid-state dosimeters are capable of recording many different types of data, including dose, dose rate, kVp levels, half-value layer, and more. The output reading is immediately available on the detector unit and can be distributed to a database via wire or Bluetooth technology. It has the ability to record doses over

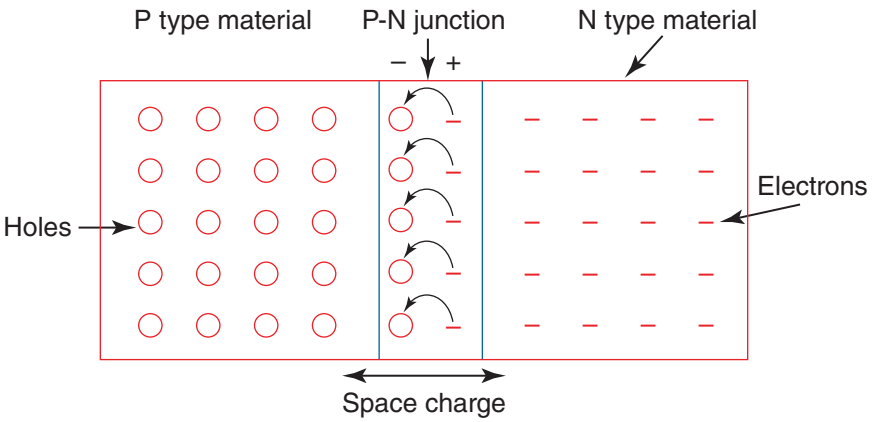


Figure 5-20 N- and p-type semiconductors in a circuit.

Figure 5-21 A solid-state dosimeter using semiconductor technology.

Figure 5-22 A RaySafe dosimetry tool.

a huge range—from 10 nGy to 1000 Gy—and is sensitive from 35–160 kVp, with units specific for mammography covering a 22–40 kVp range. Quoted uncertainties for dose readings for these meters are 5% for the standard units and 2% for mammography.

While these units are highly effective, there are three main limitations: (1) cost as these units can cost up to $10,000 or more; (2) they cannot be used in real-time imaging, as they attenuate x-rays and the unit would be visible on the image; and (3) they do not measure backscatter radiation.

Radiochromic Film

When one talks of radiation hazards within diagnostic x-ray departments, traditionally the focus has been on stochastic effects, with deterministic effects such as skin lesions being the domain of radiation therapy centers. In recent years, however, certain procedures within diagnostic radiology—particularly those including interventional components—have been changing this focus, since the prolonged exposures and high doses associated with those techniques are leading to skin burns and similar deterministic changes being reported. This means that for these examinations, skin dose monitoring requires serious consideration—in addition, of course, to the usual dose measuring that goes on. Unfortunately, the previously mentioned dosimeters have certain limitations when it comes to measuring skin doses: DAP and solid-state meters cannot be placed at the skin surface during a procedure, and while one or a number of TLDs can be positioned on the skin, movement of the patient into multiple different positions would require a high number of TLD placements for a comprehensive series of measurements. One potential solution to providing skin doses is to use clever methods of calculation to estimate skin dose from other dose recording metrics (Jones and Pasciak 2012), and previous workers have shown good to fair levels of correlation between maximum skin dose and DAP and fluoroscopic time measurements (Chida et al. 2006). While such calculations appear to be reasonable approaches for estimating a single parameter such as peak skin dose and can be done retrospectively once a procedure has been shown to be prolonged, comprehensive description of the spatial distribution of the dose across the skin surface requires an alternative strategy. This is the role of the radiochromic film.

Radiochromic film is a medium that is sensitive to radiation and will change its color depending on the level of exposure it has received, without any complex chemical processing (**Figure 5-23**). Typically, it consists of an active radio-sensitive layer around 0.03 mm thick sandwiched between two polyester layers. While its greatest use is in radiation therapy, types now available are suitable for diagnostic purposes, with the range of products

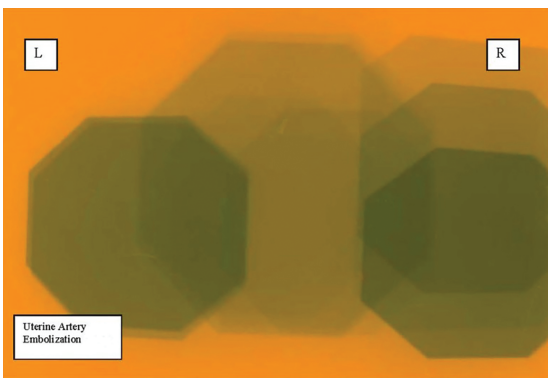


Figure 5-23 Radiochromic color change following an uterine artery embolization interventional procedure. The different colors represent different doses, which then can be profiled graphically.

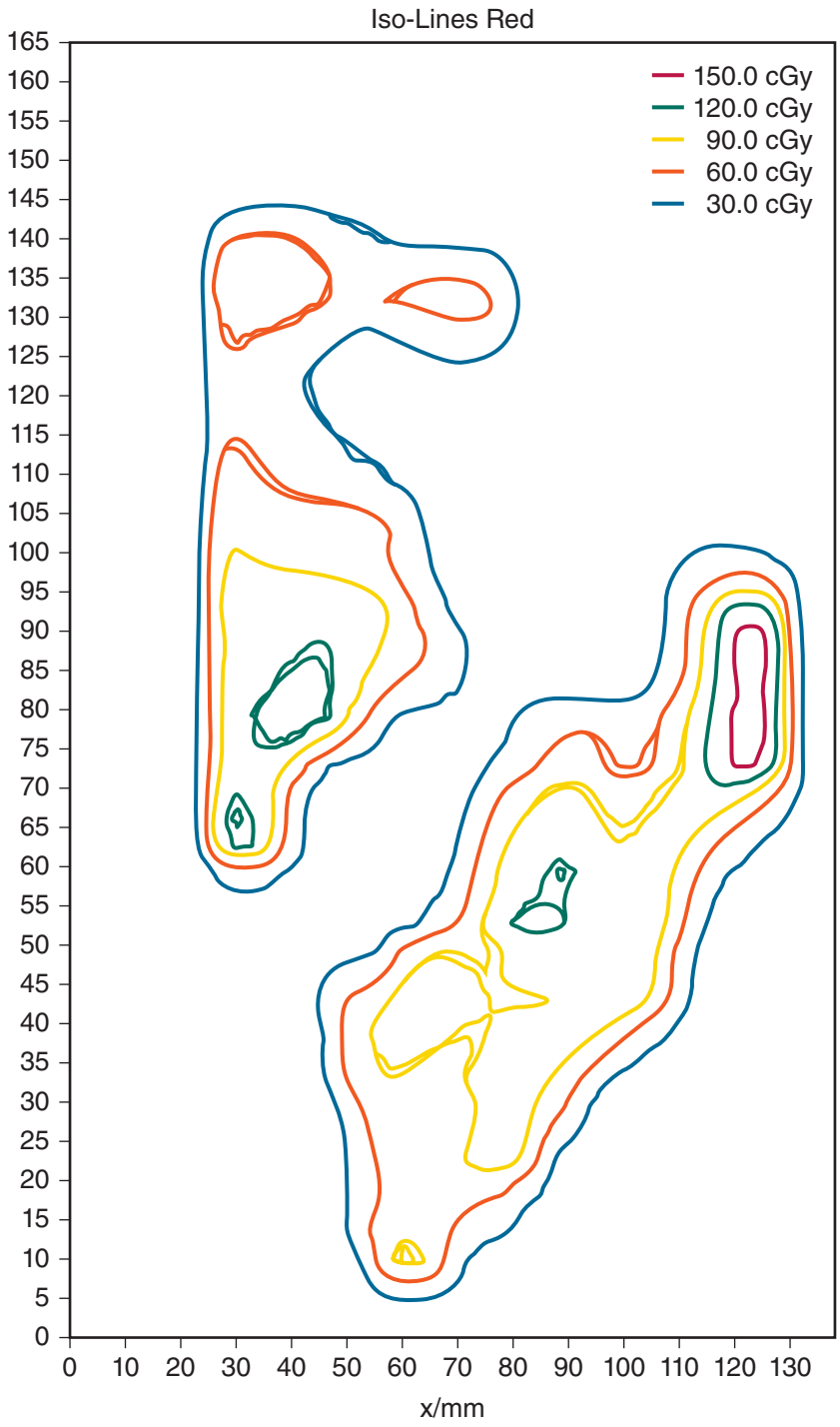


Figure 5-24 Profiling of radiation dose in two dimensions.

Courtesy of David Lewis, Ashland Specialty Ingredients (2011).

available facilitating the recording of doses from as low as 1 mGy up to 15 Gy. The spatial resolution—that is, the ability to demonstrate in fine details which part of the skin was exposed to which dose—is continually improving.

Clearly the color on the film that represents the dose must be accurately read if precise conclusions about spatial exposure distributions are to be made. Subjective opinions about the color will only go so far, and equipment such as calibrated densitometers or flatbed scanners are required for more accurate descriptions. Once scanned or measured appropriately, contour images can be produced demonstrating the specific dose distributions (**Figure 5-24**).

It is important to note that radiochromic film does have certain limitations that require careful consideration by the user. Its sensitivity to light and high temperature (>60°C), and the fading of colors post-irradiation mean that strict protocols must be adhered to if accurate measurements are to be made. Also, detailed information of the position of the film on the patient is required if information on doses deducted from the film's coloring will be precisely allocated to specific skin regions. Finally, since the film size can vary from 12.5 cm^2 × 12.5 cm^2 to 20 cm^2 × 20 cm^2, several films would need to be placed for a comprehensive measurement (which may not be the easiest thing in a complicated interventional procedure).

Summary of Key Concepts

1. **Risks associated with diagnostic imaging**. Every radiation exposure in medical imaging departments will introduce a risk of inducing a cancer, but should present a benefit to the patient as long as the exposure is justified. The responsibility of the radiographer or radiologic clinician is to ensure that the radiation dose is minimized and the benefit maximized for each examination that takes place.
2. **Justification and optimization principles**. Each x-ray exposure must be justified: it should have a good medical reason for performing it. As referred to many times within this text, irradiating individuals introduces a risk; however, this risk is acceptable if it is outweighed by the benefit that is provided diagnostically by the resultant image or images. Optimization of x-ray procedures means employing technologies or techniques that can reduce the radiation dose to patients (and staff) without sacrificing in any way the clinical information (relevant to the patient's condition) provided by performing the examination.
3. **Current radiation dose limits**. There are dose limits for workers and members of the public (e.g., family members accompanying patients) that should be adhered to. Examples of these are published by the ICRP and other national or international agencies. If these values are exceeded, appropriate measures should be put into place to minimize reoccurrence, and the exposed individual should be monitored for any adverse sequelae.
4. **Merits and applications of different dose measurement alternatives**. The principles of radiation protection—justification, optimization and dose limits (particularly the last two)—rely on effective methods on measuring radiation dose. There are a number of ways of doing this, but the main methods used in diagnostic imaging departments are thermoluminescent dosimetry; dose-area product technology; solid-state meters; and radiochromic film.

Discussion Questions

1. Discuss rationally the risks associated with radiation doses delivered during diagnostic x-ray procedures.
2. Debate the justification and optimization principles.
3. Discuss the variations in current radiation dose limits.

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