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Content:

Introduction

Diagnostic radiology has come a long way since the discovery of 'A New Kind of Rays' in Würzburg by Professor Wilhelm Conrad Röntgen in November 1895. Professor Röntgen called the unknown rays 'X rays', but they are also, most appropriately and less mysteriously, referred to as Röntgen rays after their discoverer. In the last decades, imaging techniques using X rays has developed rapidly and play an important role in modern health care.

The benefit of an X-ray examination for a patient could be assessed as how the patient's risk situation is affected. The medical condition or illness leading to the examination may imply an increased risk of deteriorating health. Correct diagnosis and proper treatment, based on the information in the X-ray image, could lower the patient's risk. For the individual the risks associated with the X-ray examination itself are small as is the radiation risk (patient absorbed dose) for well-designed imaging systems. Since the number of individuals undergoing X-ray examinations is very large, the collective absorbed dose to the whole population will be significant. Medical X-ray examinations are the man-made source giving the single largest radiation burden to the population, e.g., consisting 87% of the total radiation burden in the United Kingdom [1]. Failure to diagnose is probably the largest single risk for the patient, but for some patients adverse effects of injected contrast media may be potentially hazardous.

Ionising radiation

The nature of X rays

X rays are electromagnetic waves of the same nature as light, but with frequencies 10^4-10^5 times higher. An alternative representation is to think of X rays as individual radiation quanta or photons. Each photon has a certain amount of energy, commonly expressed in units of electron-volt, eV, (1 eV=$1.6 \cdot 10^{-19}$ J). The X-ray photons used in diagnostic radiology have energies between 10 and 150

keV (1keV=1000 eV). When X rays enter material, part of their energy is transferred to the material in interactions between the photons and the atoms. If the energy transfer is high enough to cause atomic electrons to leave the atom, the radiation is called ionising.

X-ray interactions

For X rays in the diagnostic energy range the most important interactions processes are absorptions or scatterings (figure 1). These processes are responsible for the varying transmission of photons through the patient's body which subsequently forms the image. A photon that is absorbed is lost from the beam of primary photons emerging from the X-ray tube on its way towards the image receptor. A photon that is scattered changes its direction of motion and may lose some of its energy. Information about the patient is conveyed by the primary photons; scattered photons, arising from interactions in the patient, reduce the image information content. Photon absorption and scattering in the patient result in the absorptions of energy. The energy absorbed per unit mass is called the absorbed dose and is measured in J/kg or Gray, Gy.

Absorption

In a photo-electric absorption event, a photon is wholly absorbed by one of the inner atomic electrons, which is subsequently ejected from the atom (a photo-electron). The vacancy in the electron shell is filled by an electron from an outer shell and a new photon a so called ”characteristic photon” may in some cases be created. The energy of this photon is characteristic of the atom and equals the difference in the binding energies of the two atomic electrons.

The probability for absorption varies rapidly with the atomic number of the atom, Z; increasing as $Z^{3.5}$. It decreases with increasing photon energy until the energy exceeds the binding energy of the electrons in a shell. Then more electrons can participate in the process and the probability increases considerably.

In iodine, the binding energy of the inner K-shell electrons is 33.2 keV, a value called the K-edge. The probability of X-ray absorption in iodine increases by a factor of 5 at this value (see figure 2). Iodine absorbs many more photons than soft tissue, enhancing image contrast when it is used in contrast agents.

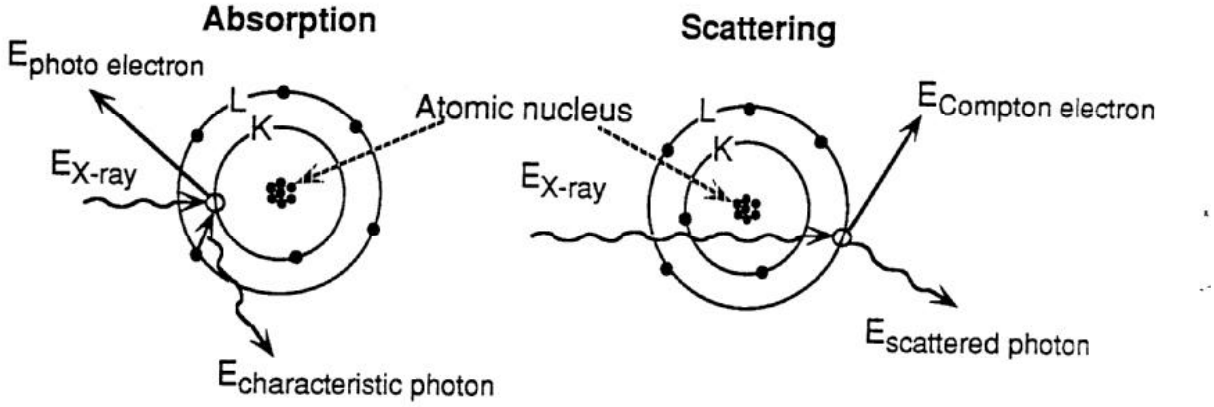

Figure 1. Schematic representation of X-ray photon absorption (a) and scattering (b) with atomic electrons. An X-ray photon with energy $E_{X\text{-ray}}$ interacts with an inner K-shell electron. A photo-electron escapes the atom and creates a vacancy in the K-shell which is filled by a L-shell electron. The difference in binding energy between the K- and L-shells may be emitted as a characteristic photon. When a photon is (Compton) scattered, the energy of the incoming X-ray is shared by the Compton electron and the scattered photon. At the photon energies used in diagnostic radiology, only a small part of the energy is transferred to the electron.

Scattering

When X-ray photons interact with atoms and scatters, the process is called Compton scattering. An incoming photon interacts with one of the outer electrons, ejecting it from the atom. Some of the energy is transferred to the (Compton) electron, the rest remaining with the scattered photon. If a 100 keV

photon is scattered 45°, its energy loss is only 5 keV. Even if a photon is scattered backwards, it cannot lose all its energy in one scattering event. This means that photons can scatter many times before their energies are so low that they are finally absorbed or escape the patient. The probability of Compton scattering is approximately independent of the atomic number and decreases only slowly with increasing photon energy.

Photons can also scatter without losing energy. This is known as Rayleigh scattering and is important only at low photon energies.

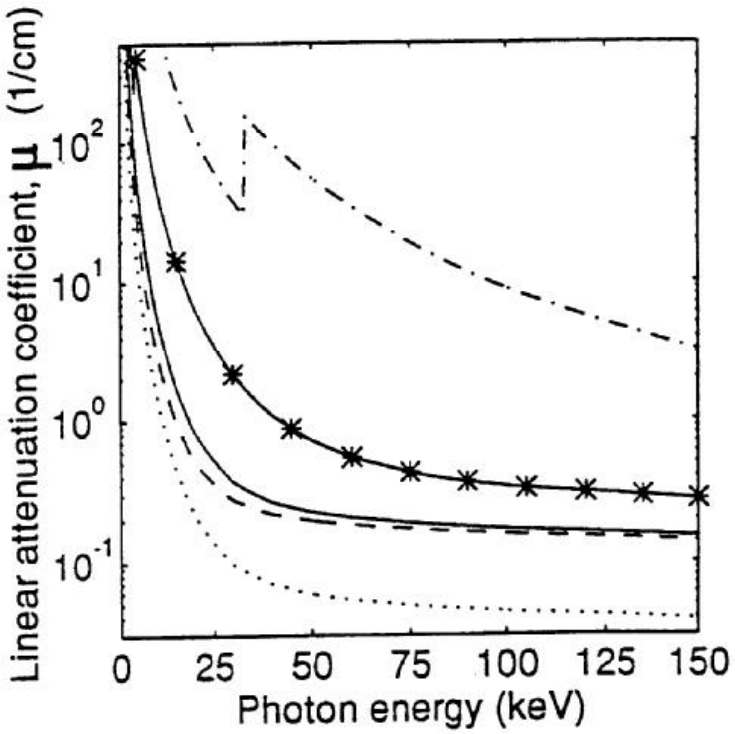

Figure 2. Linear attenuation coefficients, μ, for iodine: -.-.-; compact bone: *——*; water (soft tissue): ——; adipose tissue (fat): - - - -; lung tissue: ....., as a function of X-ray photon energy. The differences in attenuation coefficient, μ, between bone and soft tissue and soft tissue and adipose tissue decrease with increasing photon energy.

Attenuation

The number of photons of a given energy lost from a primary beam due to interaction with a thin material-layer is proportional to the thickness of the material and to the number of incident photons. The constant of proportionality is called the linear attenuation coefficient, μ. It expresses the probability per unit length that a photon of a given energy will interact somewhere during its passage through the material and it is the sum of the two interaction processes absorption and scattering. The value of μ depends on the energy of the photon and on the material (figure 2) and generally decreases with increasing energy. It is the difference in the attenuation properties of various types of tissue in the body which causes an image. When the difference is large, the tissue of interest will be distinguished more easily in the image (larger contrast). The difference in μ-values between soft tissue and bone (or iodine) is much larger than that between soft tissue and adipose tissue. Bone and iodine are thus easier to distinguish from soft tissue in the image than adipose tissue.

Imaging systems components

Figure 3 shows the main components of an imaging system including the X-ray tube, an added filter, collimator, patient, anti-scatter grid and image receptor. These will be described below. To ensure the imaging system is operating correctly, quality assurance and quality control programs are an important part of modern radiology.

X-ray tube

The photons generated in X-ray tubes are classed according to their origin as either continuous radiation (bremsstrahlung) or characteristic radiation.

In X-ray tubes, electrons emitted by heating from the negatively-charged cathode are focused and accelerated towards the positively-charged tungsten anode target by the electric field between the cathode and anode. If the potential between the anode and cathode is 100 kV, the electrons receive 100 keV kinetic energy. The broad continuous spectrum of X-ray photons (bremsstrahlung) is generated when these high-energy electrons are slowed down in the anode. Electrons rarely loose all their energy in one interaction so photons of all energies up to the maximum kinetic energy of the electron are generated, low energy photons being, however, more common. As anode material tungsten is used since it has a high melting point and high atomic number, which increases bremsstrahlung production.

When the tube potential exceeds 70 kV, K-shell electrons can escape the tungsten atom. The characteristic radiation then produced is superimposed as peaks on the spectrum of continuous radiation (see figure 4).

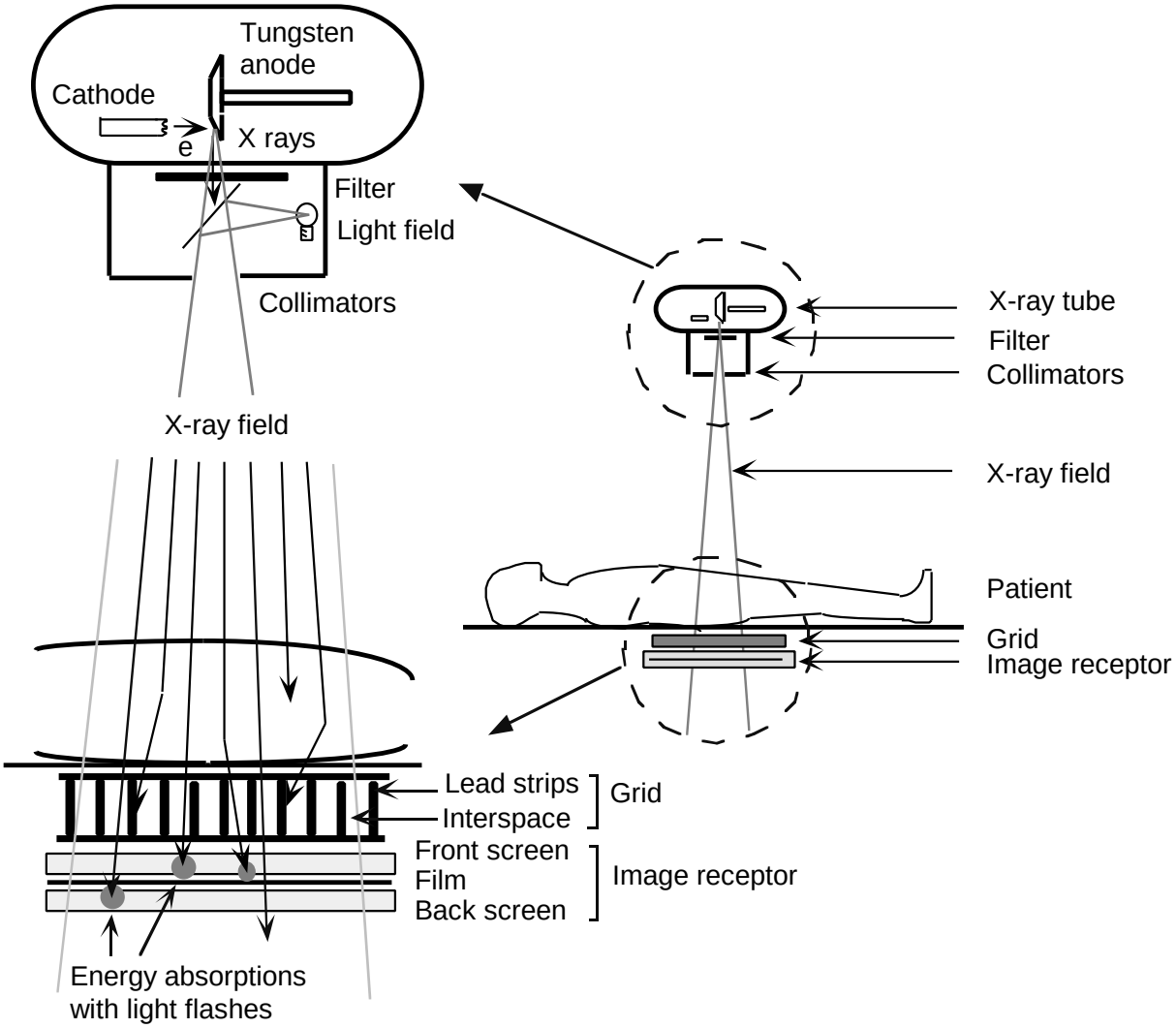

Figure 3. The components of the imaging system: X-ray tube, added filter, collimator, patient, anti-scatter grid and image receptor. Details of the X-ray tube housing and the image receptor are shown enlarged.

The parameters controlling the exposure are tube potential (kV) and tube charge (mAs), the latter being the product of tube current and exposure time. Tube potential determines the energies and fractional transmission of the photons through the body, while tube charge determines the amount of radiation generated in the tube. It should be noted that with increased tube potential

more photons will be generated per unit time since the production of X rays increases with the energy of the incoming electrons.

Filters

The tube potential applied determines the maximum energy of the X rays whereas the lowest energy of the photons reaching the patient is determined by the filtration of the beam. Beams are filtered by adding metal foils usually aluminium or copper. Since photons of low energies are highly absorbed at shallow depths in patients hence do not contribute to image formation filters should absorb low-energy photons and transmit high-energy photons.

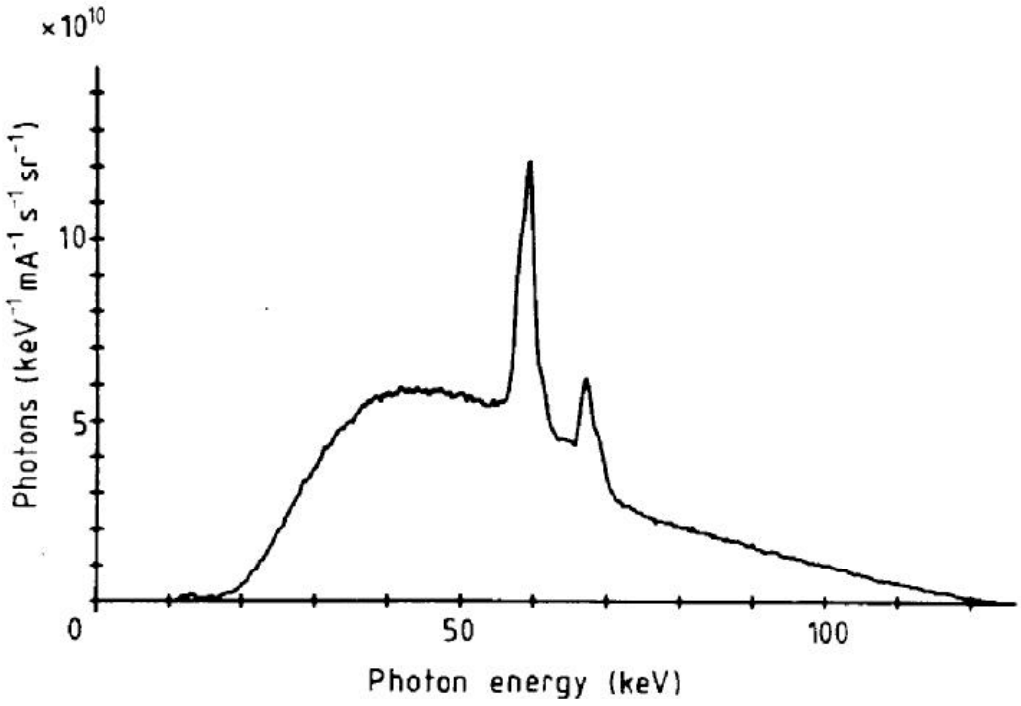

Figure 4. Measured absolute photon energy spectrum at 120 kV and 2.5 mm aluminium filtration. The characteristic photons from tungsten superimposed on the bremstrahlung spectrum were not completely resolved with this measuring technique. Reprinted with permission from [2].

The selection of appropriate thickness of the filter is a balance between image quality and absorbed dose in the patient. Due to the changes in the spectrum of photons passing through filters, average photon energies in filtered spectra are higher than in the unfiltered ones. Higher photon energies corresponds to lower μ-values. The radiation thus penetrates the patient more easily, thereby reducing the average dose in the patient needed to achieve a certain signal-level in the image receptor (film blackness). Image contrast may also be reduced so a compromise has to be reached. An X-ray spectrum for a tube potential of 120 kV with an added filtration of 2.5 mm aluminium is shown in figure 4.

Collimators

X rays are emitted in all directions from the focal spot in the X-ray tube. Collimators are therefore necessary to confine the X-ray field to the particular area of interest on the patient. Two sets of separately adjustable shutters attached to the X-ray tube determine the field. To help align the X-ray field on the patient, a light bulb and a mirror are used to form a light field projecting an image of the X-ray field on the patient.

The use of an unnecessarily large X-ray field has two disadvantages; increased patient dose and reduced image quality. The average patient dose increases if a larger portion of the patient's body is irradiated, since the energy absorbed in the patient is proportional to the area of the field. Image quality decreases since excessive amounts of scattered radiation will be generated. In good radiological practice, it is therefore essential to use as small a field size as possible.

Contrast agents

Image contrast depends on the fact that some tissues absorb more X rays than others. Since the probability for absorption increases rapidly with atomic num-

ber, contrast agents of high atomic number are used to enhance contrast. The most frequently used water-soluble contrast agents for extra-cellular space are based on iodine (Z=53) and are used to enhance the contrast of cavities, vessels and ducts. Non water-soluble contrast media containing barium (Z=56) are used to enhance the contrast of the gastrointestine canal, e.g. being used together with air in the colon.

Anti-scatter devices

If large volumes are irradiated at the same time the number of scattered photons outnumbers the primary photons at the receptor thereby reducing contrast. There are many ways to deal with this problem. One of the most efficient is to irradiate only a small part of the body at a time. This can be achieved with well-collimated beams. Another way is to use patient compression when appropriate. This technique is applied in abdominal and breast imaging (mammography) and has the further advantage of reducing the dose in the patient.

The most commonly-used method is to use an antiscatter grid. Grids consist of series of absorbing lead strips separated by transparent interspace materials such as aluminium or paper. They are similar to a Venetian blind. The lead strips are aligned so that the beam of primary photons passes through the interspaces of the grid. Scattered photons are multi-directional and most of them are absorbed by the lead strips. To work properly, grids must be precisely aligned along the X-ray beam. Otherwise too many primary photons are absorbed in the grid, which increases patient dose.

An alternative to using grids is to use an air-gap between the patient and the image receptor. With increasing air-gaps, fewer scattered photons will reach the image receptor. This method is most efficient in situations with small field sizes and thin patients, e.g., in paediatric radiology.

Image receptors

The outcome of an X-ray examination can be an analogue or a digital image. An analogue image is an image which is obtained in an analogue process, such as the exposure of a film. Ordinary film contains silver-halide which is sensitive to both light and X rays. The most frequent use of bare film as detector is in intra-oral dental X-ray examinations. By surrounding the film with two thin fluorescent screens (intensifying screens) one gets a more sensitive system and consequently lower patient dose. The energy absorbed in the screens by a single X-ray photon is converted into many hundreds of light photons that in turn expose the film. Film-blackening is then due to the light from the screens and not from direct hits of X rays. The screens are now the detector, while the film is a medium for storing and displaying the images. The active material in fluorescent screens can be made from Gd_2O_2S:Tb, LaOBr:Tm, $YTaO_4$:Nb or $CaWO_4$.

The image shows the patient anatomy with varying grey-levels: dark where X rays could easily pass through the patient (e.g., lungs) and light where this was more difficult (behind bones and contrast media). The film characteristic-curve relates the blackening on the developed film to its exposure (figure 5). A problem arises with film when the exposure setting is not properly chosen, i.e. if the image is over- or under-exposed. Image contrast is then much reduced and some of the information available in the X-ray beam behind the patient will be lost.

When dynamic images are required, image intensifier fluoroscopy or fluorography systems are used. In an image-intensifier system, X rays are absorbed in a CsI screen which emits light. The light strikes a photo-cathode that emits low energy photo-electrons. These electrons then gain energy from the electric field in the image-intensifier vacuum tube and are focused to hit the much smaller exit screen that once again converts the high-energy electrons to many light photons detected by TV- or a film-cameras.

Using subtraction techniques, one can subtract the data contained in an image obtained without contrast medium from an image of the same area obtained with contrast medium. The result is a much improved visualisation of the vessels that contained the contrast medium, all other structures being subtracted. This procedure is much facilitated by using digital techniques, in so-called digital subtraction angiography.

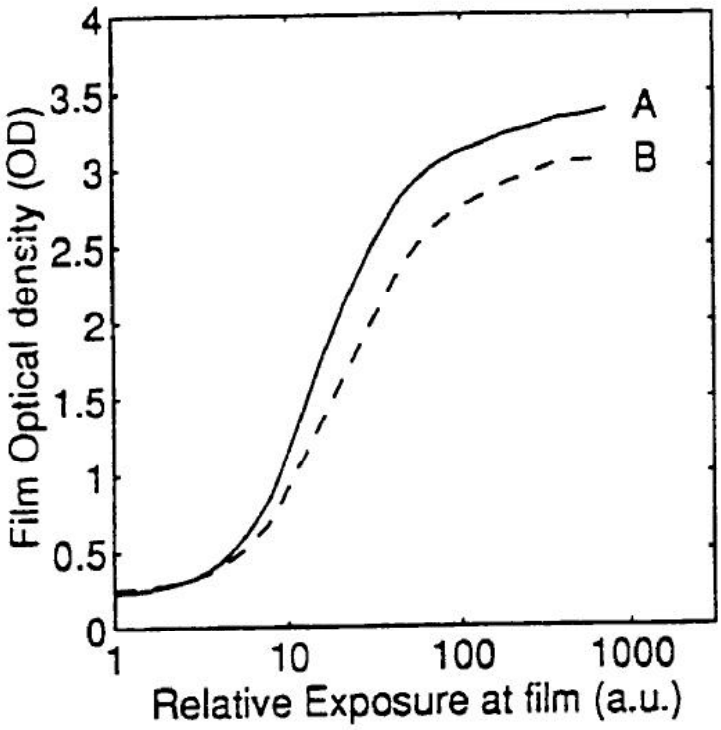

Figure 5. The characteristic curves of two X-ray films used with fluorescent screens. The curve describes the relationship between the optical density (blackening) and the relative exposure of the film. At low and high optical densities, the gradient (slope or first derivative) of the film curve is low and so is the film contrast. At optical densities between 1.0 and 2.5 the film contrast is high. The maximal film contrast for film A is higher than for film B, which on the other hand has a wider latitude, i.e., range of exposure values corresponding to a given blackening range.

Digital images

A different way of producing images is to measure values of the exposure in thousands of small areas (picture elements or pixels) in the image-plane. The number of pixels in the image differs with the application but can be 512^2, 1024^2 or 2048^2. Pixel readings are stored in a computer and can be displayed on a monitor as different shades of grey. The number of different grey-levels attainable is determined by the number of bits in each pixel, e.g. 8, 10 or 12 bits, i.e. 256, 1024 and 4096 levels of grey. All this information cannot be visualised simultaneously in one image. The storing of the images as discrete series of numbers (digits) have given this method the name digital imaging.

In digital radiography, image processing and display are separated from the process of image acquisition or detection. The advantages of digital images are many. They are rarely under- or over-exposed and several images can be created from the same set of numbers (a single exposure of the patient) by selecting different combinations of numbers and grey-levels. Image-processing techniques can easily be employed to enhance certain aspects of the images which may help interpretation. With this method, all information detected with sufficient statistical accuracy can also be displayed in the images.

Different digital image receptors have been developed, e.g., storage phosphor image plates, selenium plates and fluorescent screens in contact with a charge-coupled device (CCD). These digital image receptors are currently being developed and will constitute an even larger fraction of future radiology.

Automatic exposure control systems

To maintain the correct blackening at the region of interest in the image while changing the projection or orientation of the patient, the X-ray generator that controls the X-ray tube need continuous information about the exposure at the image receptor. By placing detectors close to the image receptor, the exposure-

level can be measured and the tube current and/or tube potential automatically controlled. The brightness of the image intensifier output screen can be measured and used by the automatic exposure control system during fluoroscopy. The brightness is kept at a constant value by altering tube potential and current.

Image quality

The benefit to the patient of the X-ray examination depends, among other things, on the quality of the image. If image quality is not sufficient for diagnosis the examination was in vain. But if image quality is higher than necessary or the imaging system is not operating properly, the dose in the patient may be unnecessary high. Optimisation of the imaging procedure thus seems appropriate. Such optimisation can be based on the radiologist's requirements of image quality for the specific task. The optimisation strategy is then to identify the method that fulfils these requirements at the lowest dose. In physical terms image quality can be expressed by the three fundamental image quality descriptors *contrast*, *sharpness (resolution)* and *noise* which are discussed below.

The information about the patient in the X-ray image is caused by differences in photon attenuation properties in different tissues which gives rise to different shades of grey in the image. Contrast, and hence thus the information in the image, is reduced by unsharpness (image blurring), by scattered radiation and by noise (stochastic variations) in the imaging system. The influence of contrast, sharpness and noise on the detectability of structures of interest depends on the imaging task, i.e., on the nature of the structure to be visualised. If it is small but has high contrast, an imaging system with high sharpness is usually preferred, but if the structure is comparatively large but with low contrast, an imaging system with low noise would be preferred.

Contrast

Contrast is the most important image quality descriptor: without contrast there is no information. Radiographic contrast depends on object contrast, receptor contrast and scattered radiation. Object contrast is the difference between the numbers of photons transmitted through the patient's body at two neighbouring area elements. It depends on differences in thickness, density, and atomic composition along rays passing the body at different positions. Obviously a thick detail of high density will stand out more clearly in the image than a small nodule with close to unit density. If the atomic number of the detail differs significantly from that of the surrounding tissues and the energy of the X-ray beam is low, the difference in the attenuation and transmission of the photons will be larger; the lower the energy the larger the difference.

If the exposure is made correctly, i.e., if the details of interest are located on the part of the film characteristic curve with the largest slope or gradient (figure 5), medical X-ray films usually enhances the object contrast. Too low or too high exposures reduce considerably receptor contrast. Film with large slopes (high gradients) enhance contrast more than films with lower slopes. On the other hand, films with lower gradients provide the opportunity to show simultaneously areas of very different attenuation or exposure levels. In digital imaging systems, contrast can be manipulated after the exposure to enhance visibility of particular details of interest. This is useful provided the noise level is not too high.

Sharpness

Sharpness (or spatial resolution) is the ability of the imaging system to depict a sharp edge. Small details are more easily detected with an imaging system with high sharpness. Different aspects of the imaging system contribute to lowered sharpness. These are geometric, object and receptor unsharpness. Geometric unsharpness is caused by the finite size of the X-ray focal spot and can be

minimised by keeping this spot small and the details to be imaged as close to the receptor as possible. Patient motion during exposure contributes to object unsharpness and can be minimised by using short exposure times. Receptor unsharpness is mainly caused by lateral diffusion of the information carriers, such as the light photons in fluorescent screens. It can be reduced by using thin screens, but this will reduce their ability to absorb X rays. The search for materials that efficiently absorb the photons but can be made thin to reduce unsharpness is important. Lenses and TV-cameras in image intensifiers are also a source of unsharpness.

Noise

Noise can be defined as variations in the image that do not correspond to variations in the patient's anatomy, but arise from stochastic processes and imperfections in the imaging chain. The emission of X rays, their attenuation in the patient and their absorption in the receptor are stochastic processes. If we repeatedly irradiate an object in what we think are identical ways, we find that the image is not exactly the same every time. The image will have an irregular grainy appearance. These stochastic small-area variations are called noise. Quantum noise contributes most to the total noise in well-designed imaging systems, and arise from the limited number of X-ray photons used to build up the image. The larger the number of X-ray photons absorbed in the receptor per unit area, the lower the quantum noise. Quantum noise can thus be reduced by increasing the irradiation (increase mAs), but the dose in the patient will then increase. A schematic illustration of the influence of noise on the detecta-bility is given in figure 6.

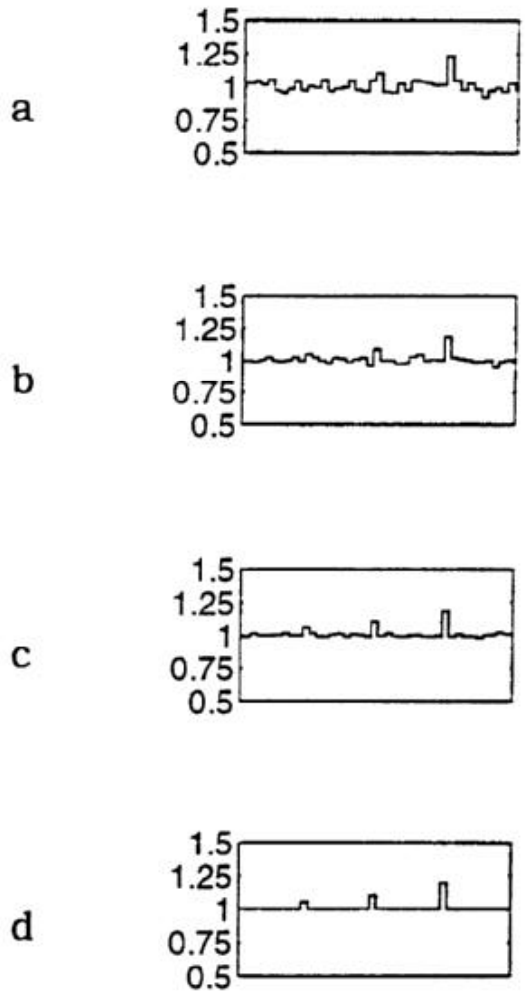

Figure 6. A simple illustration of the influence of noise on the detection of small details. Figure (d) show the details without noise, the contrasts (the signal) being 5%, 10% and 20% for the leftmost, middle and rightmost details respectively. In figures (a-c) noise is added. The standard deviation in the noise distribution decreases going from (a) to (c). The detectability of the details in the background noise, as quantified by the signal-to-noise ratio (SNR), increases from left to right in each figure, and for each contrast detail, from (a) to (c). When SNR˜5, the details are just about detectable (rightmost detail in 6a, middle detail in 6b, and leftmost detail in 6c). When SNR<5, one is not able to say with any confidence whether or not the detail is present. When SNR>5, the opportunity for detecting details correctly increases.

Figure 6d shows three details with 5%, 10%, and 20% contrast with no noise or background variation. In figures 6a-c noise is added and details of lower contrast become more difficult to detect. The noise decrease by a factor of two from fig. 6a to 6b and from 6b to 6c and details with higher contrasts can more easily be separated from the noisy background (see also chapter 3, figure 6a-c).

A measure of the accuracy of the information or of the detectability of details of interest in the image is the ratio between the signal (the contrast) and the noise. This quotient, the signal-to-noise ratio, SNR, needs to be sufficiently high for the radiologist to be able to separate (detect) low-contrast details from noisy backgrounds. For well-designed digital imaging systems, SNR increases with increasing irradiation of the patient. To double SNR, patient irradiation must be increased four times, other things being equal.

Radiation risk

Radiological protection principles

Acute effects of radiation damage was reported soon after the discovery of X rays in 1895. From the point of view of radiological protection, the main concern today is the increased risk for stochastic effects such as cancer development following irradiation with absorbed doses too low to cause acute effects. The reduction of stochastic effects is the basis principle of radiological protection [3]. The first principle states that all use of ionising radiation should be *justified*. The remitting physician must be reasonably sure that the information gained from the X-ray examination could not be found by an alternative procedure without ionising radiation. The second principle states that all exposures should be as low as reasonably achievable. It is thus necessary to consider how best to *optimise* the examination, i.e., to gain the information at the lowest patient dose. The third principle *limits doses* in individuals. Dose limits generally

do not apply to patients who should directly benefit from the examination, but are aimed at employees.

Radiation damages

Stochastic damages can be either hereditary or carcinogenic and have no threshold, i.e., the character of the damage (e.g., cancer) does not depend on the absorbed dose. The basic assumption in radiological protection is that the probability (or risk) for developing cancer later in life due to exposure increases linearly with increasing absorbed dose. This is an approximation. The radiosensitivity varies among individuals and the links between dose and risk for radiation-induced cancer at very low doses may be questioned. Lack of knowledge has led to a conservative approach, not to presuppose the existence of threshold doses below which the risk is zero. On the other hand, acute damage (e.g., skin erythema and cataracts of the eye lens), occurs only after a threshold dose has been exceeded.

Increased frequencies of occurrence of leukaemia and cancer have been noted among the survivors of the nuclear bombing in Japan. These people were exposed to almost homogeneous total-body irradiation. By studying the over-frequency of cancer in this group, it has been possible to derive relative risk-factors or tissue-weighting factors for different organs.

Acute damage such as skin erythema has till now been very rare. Recently, however, in complicated interventional procedures with long time fluoroscopy at high dose rates, skin erythema has been reported.

Dosimetric concepts

In an X-ray examination, the patient is not homogeneously exposed to radiation. The dose decreases rapidly with depth in the patient's body and even more rapidly outside the boundaries of the radiation field. The effective dose was defined [3] to express the stochastic risk of cancer induction and genetic injury in

such circumstances. It is the product of the average organ or tissue dose and the relative tissue-weighting factor for that organ or tissue type, summed over all exposed organs in the body. The effective dose is measured in units of Sievert, Sv. The organs with the largest tissue-weighting factors are gonads, red bone marrow, colon, lung and stomach. The effective dose per examination, the doses in the uterus and ovaries are shown in table 1. Reference doses for good practice for some common X-ray examinations have been suggested along with image quality criteria and image technique settings [4]. If the patient dose at your clinic is significantly larger than the reference dose level, it is an indication that your imaging system or technique is not operating in an optimal way and needs revision.

The risk for developing fatal cancer after exposure to ionising radiation has been estimated to be 5%/Sv or 0.005%/mSv. It should be noted that this risk is 2-4 times larger for children. In comparison, the natural background effective dose in Sweden, excluding exposure to radon, is 1 mSv. It is very difficult to compare risks, but to put it into some kind of perspective, the risk associated with a dose of 1 mSv is about the same as smoking 60 cigarettes or driving a car 5000 km. It is not uncommon to underestimate risks in every-day activities while overestimating risks in rare and unknown activities.

Table 1. Effective doses and absorbed doses in the uterus and the ovaries for some common X-ray examinations. The data originate from a survey of 20 English hospitals [5] and are averages from the patient sample measured. The effective doses has been corrected for the new tissue weighting factors [3].

Examination	Effective dose mSv	Uterus dose mGy	Ovary dose mGy
Chest	0.04	*	*
Head	0.12	*	*
Pelvis	0.9	1.7	1.2
Abdomen	1.4	2.9	2.2
Lumbar Spine	2.0	3.5	4.3

* Absorbed dose is less than 0.01 mGy

Optimisation of image quality and dose

An optimisation strategy can be thought of as a procedure that determines the imaging technique that utilises the information most efficiently, in other words, by searching for imaging techniques maximising the quotient of the signal-to-noise ratio (SNR) to the radiation risk (effective dose). It is then up to the radiologist to decide on the level of image quality required. For example, the choice of optimal energy spectrum (tube potential) is a compromise between the needs for low dose, which requires a high tube potential, and high contrast, which requires a low tube potential. Knowledge of the variations of dose and contrast with tube potential enables the radiologist to make a sensible choice.

To take fully advantage of the information in images requires digital image capture and possibilities of image processing. This is facilitated by separating

image capture and image display. Computerised tomography (see chapter 3), which was introduced in the seventies, uses digital image capture and has thus a good ability to register even small differences in attenuation (or contrast). A danger with digital techniques is that in principle all the information present can be captured. Very small details and very small tissue differences can be detected provided the irradiation of the patient is increased. With the increasing capacity of modern X-ray tubes and even smaller pixel sizes, we could well come to a point where demands for more and more information (higher and higher image quality) will lead to unacceptably high doses. More efficient utilisation of the X rays could counteract this scenario.

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