5.7.1. Quantities describing X ray output

Total photon fluence is not a satisfactory quantity to describe X ray output; rather, it is the spectral distribution of the photon fluence as a function of photon energy that is useful for research in X ray imaging. Spectral data are rarely available for individual X ray units, although computer programs exist that give useful simulations.

X ray tube output can be expressed in terms of the air kerma and measured free in air (see Chapter 22). A measure of the penetration and the quality of the X ray spectrum is the HVL. The HVL is the thickness of absorber needed to attenuate the X ray beam incident air kerma by a factor of two. In diagnostic radiology, aluminium is commonly chosen as the absorber, giving the HVL (unit: mm Al).

5.7.2. Tube voltage and current

Figure 5.18 shows the effect of tube voltage on spectral distribution. Both maximum and mean photon energy depend on the voltage (kV). The shape of the low energy end of the spectrum is determined by the anode angle and the total filtration. Note the appearance of characteristic radiation in the 100 kV beam and the increase in photon yield with increasing tube voltage. Tube current has no

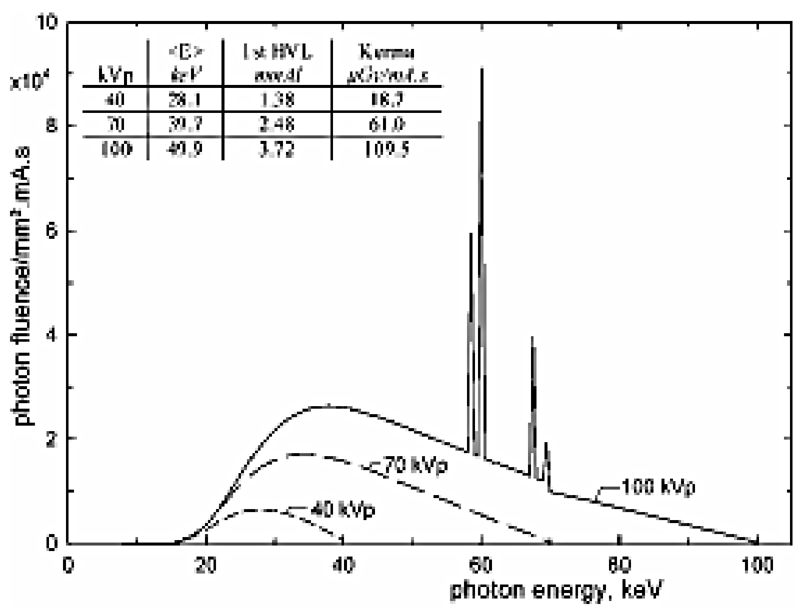

FIG. 5.18. X ray spectra for various tube voltages and a tungsten target (constant voltage, anode angle 12°).

5.7.3. Tube voltage ripple

Figure 5.19 shows spectral variations for a tube voltage of 70 kV for various voltage ripples. A DC voltage gives the hardest spectrum with maximum photon yield. With an increase in ripple, the yield drops and the spectrum softens.

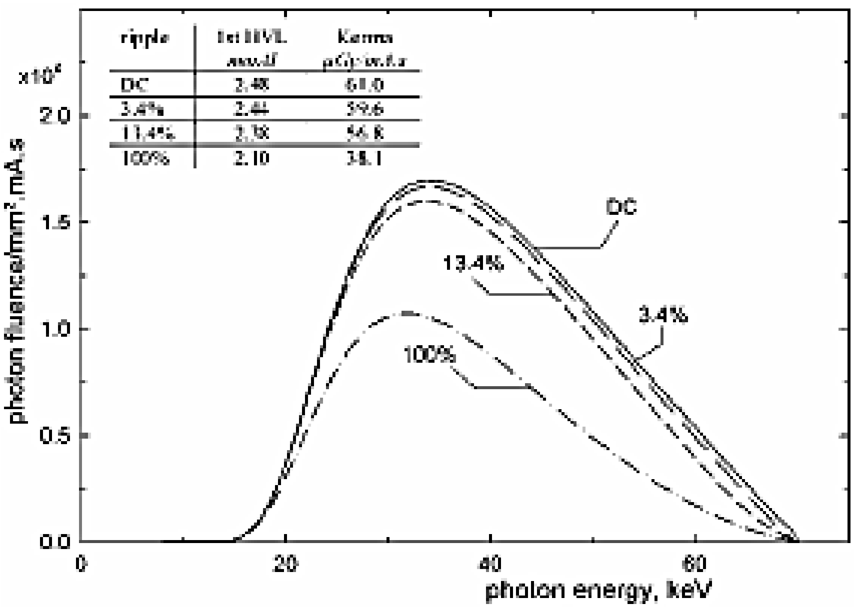

FIG. 5.19. Variation of X ray spectra from a tungsten target with tube voltage ripple at 70 kVp tube voltage. DC: constant potential; 3.4%: 12-pulse or converter generator; 13.4%: 6-pulse generator; 100%: 2-pulse generator.

5.7.4. Anode angle

The anode angle determines the degree of X ray absorption in the anode material. A decrease in anode angle causes an increase in the absorption length within the target. Accordingly, the maximum photon energy remains unchanged, but hardness increases and yield drops with decreasing anode angle (Fig. 5.20).

5.8. FILTRATION

As low energy photons do not contribute to the formation of an image, filters are used to reduce the low energy component. Figure 5.21 illustrates the effect of added filters on an X ray spectrum (90 kV, 3.4% ripple). Again,

Chapter 6

PROJECTION RADIOGRAPHY

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6.1. INTRODUCTION

In its simplest form, X ray imaging is the collection of attenuation shadows that are projected from an ideal X ray point source on to an image receptor. This simple form is true for all X ray imaging modalities, including complex ones that involve source and receptor movement, such as computed tomography (CT). This simplified view, however, is made vastly more complex by the non-ideal point source, by the consequences of projecting a 3-D object on to a 2-D detector and by the presence of scattered radiation, generated within the patient, which will degrade any image that is captured.

6.2. X RAY IMAGE FORMATION

6.2.1. Components of an imaging system

The principal components of a system for X ray projection radiography are illustrated in Fig. 6.1. Of these components, the grid and the automatic exposure control (AEC) are optional, depending on the imaging task. Further components such as shaped filtration, compression devices or restraining devices may be added for special cases. The X ray tube and collimation device are described in Chapter 5 and the image receptor systems are described in Chapter 7.

When considering such systems, the concept of an ideal imaging task is often useful, as illustrated in Fig. 6.1. These concepts are covered in detail in Chapter 4 and are discussed only briefly here. When considering the ideal imaging task — the detection of a detail against a uniform background — the ideal X ray spectrum is monochromatic when the three constraints of patient dose, image quality and X ray tube loading are considered. Any particular projection may consist of more than one such task, each with a different ideal monochromatic energy. The choice of X ray spectrum for each task is, therefore, always a compromise, so that the actual bremsstrahlung and characteristic radiation spectrum provide the best approximation to the ideal monochromatic spectrum for the particular projection.

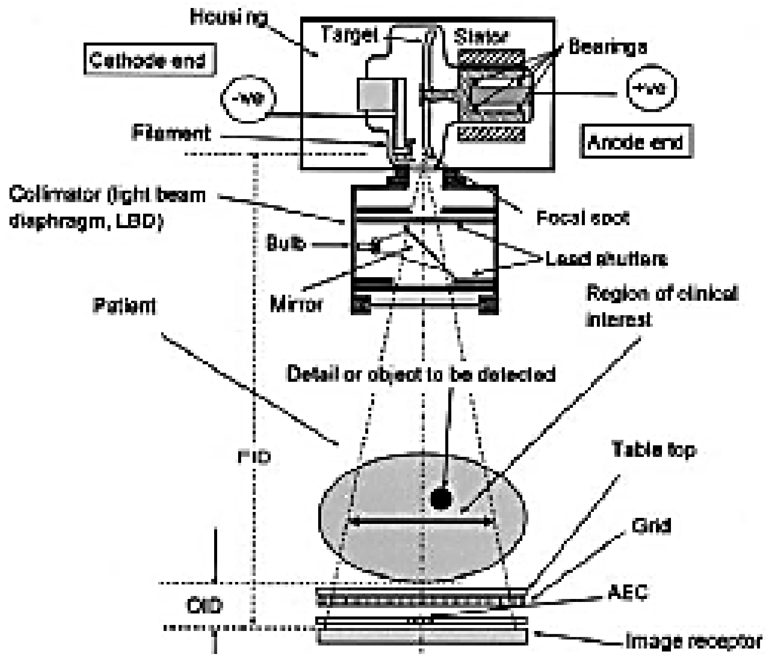

FIG. 6.1. Components of a projection radiography system, including an ideal imaging task, for the detection of a detail against a background. FID: focus to image distance; OID: object to image distance.

Considering an ideal imaging task, as illustrated in Fig. 6.1, contrast may be defined simply as $C = \Delta B/B$, where B is the image brightness (or shade of grey) in a background region and ΔB is the difference in brightness for a small detail. For small values of ΔB, linearity of brightness with X ray intensity (I) is assumed, so the contrast is $\Delta I/I$. This is generally valid for a particular monochromatic spectrum. For a real polychromatic spectrum, a monochromatic spectrum with the average energy of the actual spectrum may be used as an approximation, or the result can be integrated over all spectral energies. Since the X ray intensity is related to thickness by the attenuation law, it follows that the primary contrast for a detail of thickness x_d and linear attenuation coefficient μ_d embedded in a material of linear attenuation coefficient μ_b is given by:

$$C_P = 1 - e^{-(\mu_d - \mu_b)x_d} \qquad (6.1)$$

To find the average contrast for a particular detail, Eq. (6.1) should be integrated over the detail. For thin spherical details (e.g. microcalcifications in mammography, solitary pulmonary nodules in chest X rays), this is straightforward and results in a contrast that is 2/3 the contrast obtained for a ray passing through the centre of the detail.

With regard to the relationship between the linear attenuation coefficient and the mass attenuation coefficient (see Section 2.3.3), contrast will exist for details that differ in mass attenuation coefficient, or in density, or both. The contrast will depend on the thickness of the detail, but not the thickness of surrounding tissue. Since the values of μ reduce as photon energy increases, the contrast is seen to be inversely related to the kV setting. Thus, kV may be considered to be the contrast control, where contrast is strictly the detail contrast. For screen film imaging, the difference in optical density (OD) due to the detail is proportional to the subject contrast multiplied by the gamma of the screen film system (see Section 7.3.4). For digital image receptors, the relationship is more complex, since the contrast of the displayed image is independently adjustable.

If the energy absorbed in a small region of the image receptor due to primary rays is E_p, and that due to secondary rays is E_s, then scatter may be quantified by the scatter fraction:

$$\mathrm{SF} = E_s/(E_p + E_s) \tag{6.2}$$

or by the scatter to primary ratio:

$$\mathrm{SPR} = E_s/E_p \tag{6.3}$$

The relationship between the two is:

$$\mathrm{SF} = ((\mathrm{SPR}^{-1}) + 1)^{-1} \tag{6.4}$$

In the presence of scattered radiation, Eq. (6.1) becomes:

$$C_p = 1 - e^{-(\mu_d - \mu_b)x_d} \frac{1}{1+\mathrm{SPR}} \tag{6.5}$$

Clearly, minimization of scatter is important, leading to the use of antiscatter techniques (see Sections 6.3.4 and 6.3.5). Accurate collimation to the region of clinical interest also minimizes scatter, as well as reducing the dose to the patient.

6.2.2. Geometry of projection radiography