THE PHYSICS OF MEDICAL IMAGING

Edited by

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2.5 X-RAY TUBES

2.5.1 Tube construction

The x-ray tube used in diagnostic radiology consists of an oil-filled housing containing an insert (Forster 1985 pp56–68), which is an evacuated envelope of heat-resistant borosilicate glass within which are mounted a filament and an anode (figure 2.8). The filament is heated by passage of an electric current. It produces a narrow beam of electrons, which are accelerated by a potential difference of between 25 and

150 kV and strike the anode. The electrons interact with the material of the anode, slow down and stop. Most of the energy absorbed from the electrons appears in the form of heat, but a small amount (less than 1%) appears in the form of x-rays. Some of these x-rays pass through the exit windows of the insert and housing and through the patient to form the x-ray image. X-rays that are emitted in other directions are absorbed by the housing. The complete tube assembly is mounted on a support and provided with collimation so that the beam direction and the size of the radiation field may be varied as necessary.
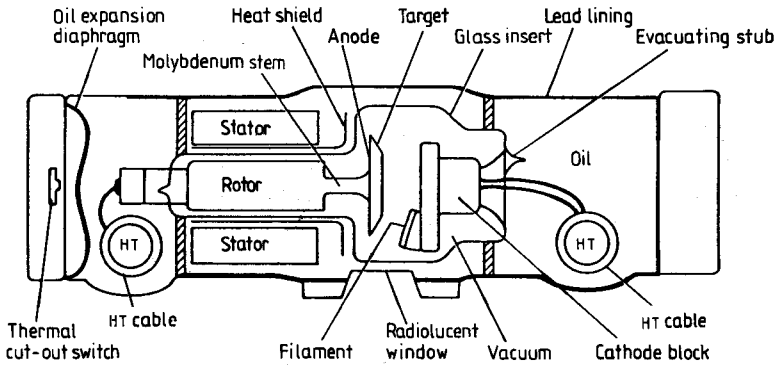

Figure 2.8 The construction of a rotating-anode x-ray tube. (Reproduced with permission of MTP Press Ltd from Forster (1985).)

The design of the filament and the electron optics that guide the electrons to the anode is very important because the unsharpness in the image may be limited by the size of the x-ray source, and the output from the tube is determined by the electron current striking the anode. The filament is constructed from a spiral of tungsten wire (melting point 3410 °C), which is set in a nickel block. This block supports the filament and is shaped to create an electric field that focuses the electrons into a slit beam. The anode has a bevelled edge, which is at a steep angle to the direction of the electron beam. The exit window accepts x-rays that are approximately at right angles to the electron beam so that the x-ray source as viewed from the receptor appears to be approximately square even though the electron beam impinging on the target is slit-shaped (figure 2.9). The choice of the anode angle θ will depend upon the application, with the angle being varied according to the requirements of field and focal spot sizes and tube output. For general-purpose units, an angle of about 17° is appropriate. In many cases the anode disc will have two bevels at different angles and two filaments, so that either a fine or a broad focus may be selected.

It has already been observed that most of the energy in the electron beam is deposited in the target in the form of heat, and one of the

problems faced by the tube designer is how to limit the heat deposited in the target area and how to remove it from that area as quickly as possible. The use of a slit source of electrons helps by spreading out the target area and this idea can be extended by using a rotating anode, so that the electron beam impinges on the bevelled edge of a rotating disc and the target area is spread out over the periphery of the disc. A rotation speed of about 3000 RPM and an anode diameter of 10 cm are used in general-purpose units. It is also possible to use a stationary target and these are sometimes to be found in simple, low-power equipment, such as mobile and dental x-ray units.
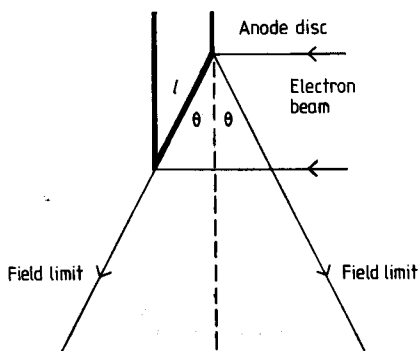

Figure 2.9 The use of a bevelled anode to reduce the effective focal-spot size. The width of the electron beam is $l \cos \theta$ whereas the focal-spot size as viewed from the central axis of the x-ray field is $l \sin \theta$.

The anode is usually constructed from tungsten although molybdenum is used for special applications where a low-energy x-ray beam is required (see §2.5.2). Tungsten has an atomic number of 74, acceptable thermal conductivity and thermal capacity, and a high melting point. The high atomic number is important because the bremsstrahlung yield from the target increases with atomic number and the x-ray spectrum from an element of high atomic number is well suited to imaging the thicker body sections. Improved tube lifetime can be obtained by using a 90/10 tungsten/rhenium alloy. This reduces crazing of the anode surface caused by the continual heating and cooling processes to which it is subjected. It is important that the anode disc has a high thermal capacity. A larger anode disc will give a higher rating and a shorter exposure time, and the greater thermal capacity associated with an increase in anode volume will allow the possibility of a shorter time interval between exposures. For heavier-duty applications, the thermal capacity can be increased by using a molybdenum backing to the anode. Molybdenum has a higher specific heat than tungsten (table 2.4) and the heat capacity for an anode of this type would typically be 250 000 J.

The anode disc is mounted on a thin molybdenum stem. This reduces heat flow backwards and prevents the rotor bearings, which are made from copper, from overheating. The heat loss from the rotating anode is mainly radiative. The requirements of x-ray tubes for CT are mentioned in Chapter 4.

2.5.2 X-ray spectra

The shape of the x-ray spectrum will depend upon the target material, the potential and waveform applied to the tube and the effects of any filters placed in the x-ray beam. Table 2.4 shows the properties of the two most commonly used target materials and figures 2.10, 2.11 and 2.12 show three typical x-ray spectra. The first two spectra are for tubes with tungsten targets and the other spectrum is for a tube with a molybdenum target. Also shown are the spectra after modification by transmission through the body.

Table 2.4 Properties of molybdenum and tungsten.

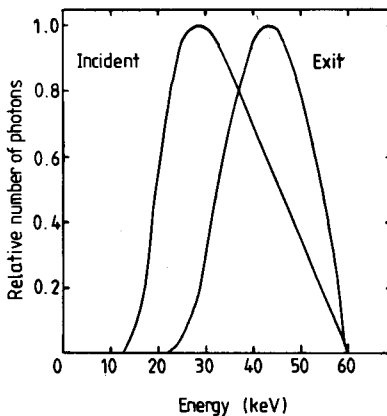|	Mo	W
Atomic number	42	74
K x-ray energies (keV)	17.4–19.8	58.0–67.7
Relative density	10.2	19.3
Melting point (°C)	2617	3410
Specific heat (J kg^{-1} °C^{-1})	250	125

Figure 2.10 X-ray spectra for an x-ray tube with a tungsten target; 60 kV constant potential with 1.5 mm aluminium added. The spectra are shown both before and after attenuation by 9.5 cm soft tissue plus 0.5 cm bone. (The spectra are based on the work of Birch *et al* (1979).)
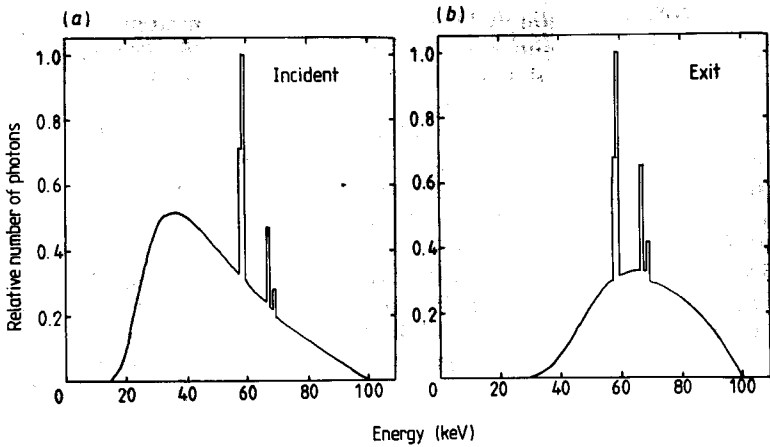
Figure 2.11 X-ray spectra for an x-ray tube with a tungsten target; 100 kV constant potential with 2.5 mm aluminium added. The spectra are shown both before and after attenuation by 18.5 cm soft tissue plus 1.5 cm bone. (The spectra are based on the work of Birch *et al* (1979).)
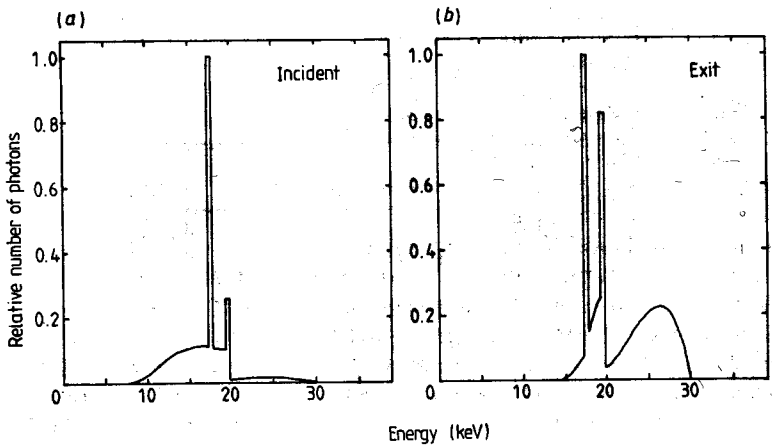

Figure 2.12 X-ray spectra for an x-ray tube with a molybdenum target; 30 kV constant potential with 0.03 mm molybdenum filter. The spectra are shown both before and after attenuation by 5 cm tissue. (The spectra are based on the work of Birch *et al* (1979).)

The tungsten spectra are well suited to imaging the thicker body sections because of the energies of the tungsten characteristic x-rays. Molybdenum has lower-energy x-rays, which are more appropriate for imaging thinner body sections at high contrast. It is used in x-ray units specifically designed for mammography. In figure 2.12 the x-ray

spectrum from the molybdenum target has been modified by the use of a molybdenum filter, which heavily attenuates any x-rays that have energies above its K-edge of 20.0 keV. This is standard practice for a mammographic x-ray unit.

There is a large difference between the x-ray spectra before and after passage through the patient. The difference between these spectra is due to the photons that interact in the patient and deliver dose. If the spectrum is too soft, then the lower-energy photons will only contribute to patient dose and not to image contrast, and it is important that they be removed from the x-ray beam before it reaches the patient. This can be achieved by the use of aluminium or copper filters, depending on the tube potential.

The choice of the tube potential and filtration most appropriate to a particular application will be a compromise between contrast and dose, as explained in §2.4. As we have seen already, we cannot reduce the dose to a very low level because of the noise in the image and it is possible to use this fact to predict optimum energies for imaging different body thicknesses. Dance and Day (1981) have modelled the mammographic examination and have investigated how the signal-to-noise ratio in the image varies with photon energy and breast thickness. This is a particularly important investigation to optimise because it is at present the technique of choice for population screening for breast cancer and, even though dose levels are low, there is a small risk of carcinogenesis associated with the examination. (Comments on the relation between the roles of x-ray mammography and other breast imaging modalities are to be found in Chapters 10 and 11. The question of risk is addressed in Chapter 15.) Sample results from Dance and Day (1981) are shown in figure 2.13. Each curve is for fixed breast thickness and gives the signal-to-noise ratio for imaging a 0.1 mm microcalcification (a speck of calcified material that is often found within the breast and which it is important to be able to see on an x-ray) against a uniform background. The breast dose is fixed at 1 mGy. Each curve passes through a maximum and the position of this maximum gives the optimum energy for imaging a microcalcification for a breast of a particular thickness. At this energy, the exposure could be reduced to achieve the desired signal-to-noise ratio at the least possible patient dose. The curves shown in the figure are appropriate to imaging systems where acceptable results can be obtained over a wide range of exposures, so that the dose can be adjusted to give the required noise level without the image being either under- or overexposed. Digital imaging systems satisfy this requirement but the more conventional film-based systems may not. Nevertheless, the curves shown in figure 2.13 provide useful data, which can be of value in clinical practice. They demonstrate clearly how the optimum energy varies with breast thickness. If we look at the mammographic x-ray spectrum shown in figure 2.12, we see that it is most appropriate for imaging the smaller breast. We can also see that it is less appropriate for imaging very large breasts where it would be preferable to use a slightly higher-energy spectrum. Such a spectrum

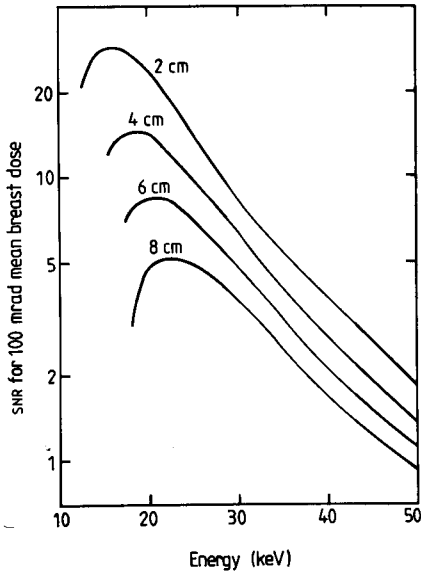can be obtained by using a different K-edge filter and a tungsten target.

Figure 2.13 Signal-to-noise ratio for a 100 μm microcalcification (calcium hydroxyapatite, $Ca_5(PO_4)_3OH$) embedded in a breast. The SNR has been calculated for a Kodak Min-R screen and a mean breast dose of 1 mGy. Curves are given for breast thicknesses in the range 2–8 cm. (Reproduced from Dance and Day (1981) with permission of the Office for Official Publications of the European Communities.)