5.4.2.4. Capacitor discharge generators

In places with inadequate mains supply, or in remote locations, capacitor discharge generators are helpful. A capacitor is charged to a high voltage just before an exposure. The capacitor is connected to the X ray tube with the start and length of exposure controlled by a grid. High tube currents and short exposure times can be obtained. However, discharging a capacitor implies a falling tube voltage during exposure. Typically, voltage drops of ~1 kV/mAs are usual. As kerma drops with voltage, the appropriate exposure of thick body parts can be problematic.

5.4.2.5. Constant voltage generators

Constant voltage generators achieve a DC high voltage with minimal ripple through the use of a closed loop linear voltage controller (e.g. high voltage triodes) in series with the tube. High frame rates and voltage stability are achieved. Constant potential generators use a complex technology with high costs of investment and operation and, as a consequence, have lost popularity.

5.4.2.6. Comparison of generator technologies

Figure 5.14 shows a comparison of voltage waveforms together with their associated kerma output. In radiology, it is desirable to keep exposure times as low as achievable. One-pulse waveforms produce radiation in only half of a cycle, and double the exposure time compared with 2-pulse voltages. As the kerma output rises approximately with the square of the tube voltage, there is a substantial amount of time in a half wave of 1- and 2-pulse waveforms, with little or no contribution to kerma output, again effectively increasing the exposure time.

The 1- and 2-pulse waveforms also yield softer X ray spectra, which implies increased radiation dose to the patient. Both the exposure time and the patient dose indicate that the optimum waveform would be a DC voltage with essentially no ripple, but 12-pulse and high frequency generators are near optimum.

Generators transforming mains AC voltages suffer from external voltage instabilities. Devices for compensating for these fluctuations are often integrated into the generator design; high frequency generators that provide tube supplies with higher stability and accuracy are currently the state of the art.

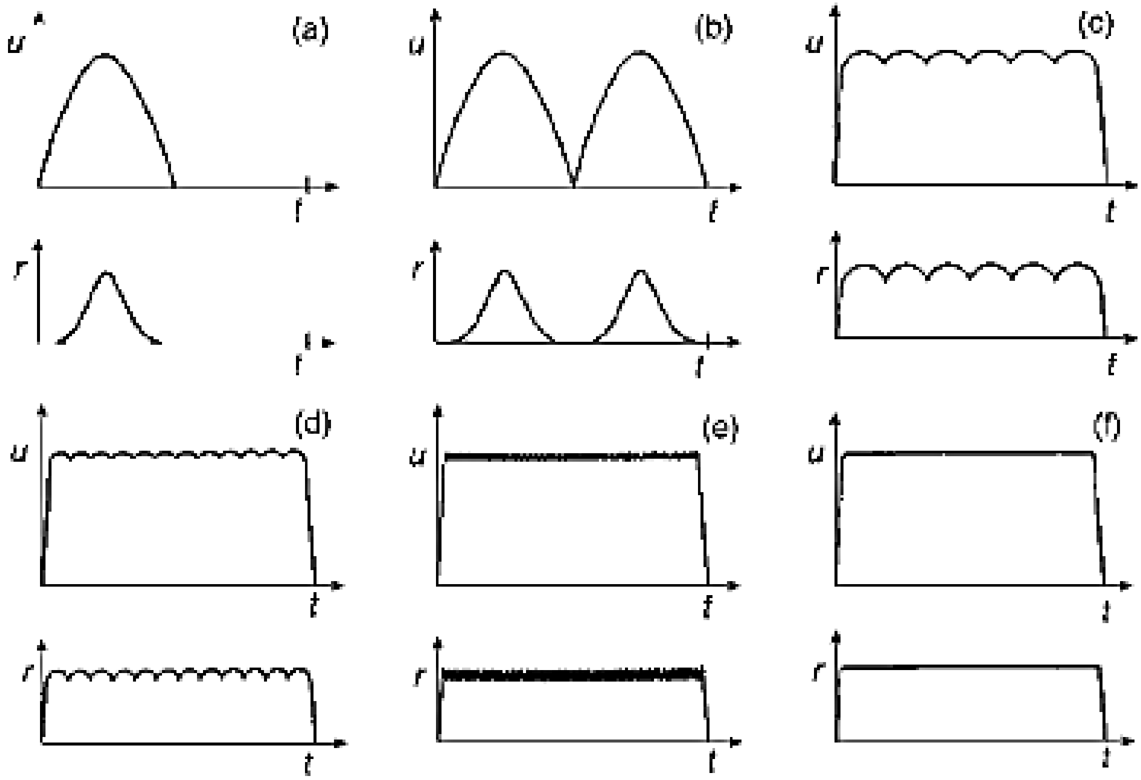

FIG. 5.14. Voltage waveforms (u) and associated tube output (dose rate (r)) versus time (t) for (a) 1-pulse, (b) 2-pulse, (c) 6-pulse, (d) 12-pulse, (e) high frequency and (f) constant voltage generators.

5.4.3. Exposure timing

Exposure of a radiograph can be set manually by choosing tube current and exposure time. Except in examinations with little variability in body dimensions (e.g. extremities), an AEC is mandatory to achieve a consistent image quality or film density. The AEC terminates an exposure when the image receptor has received a preset level of radiation.

The AEC system consists of one to three radiation detectors (ionization chambers or solid state detectors). The signal from these detectors is amplified and integrated, corrected for response in photon energy and dose rate, and finally, compared with the preset dose level. The exposure is terminated when the chosen level is attained. In case the AEC does not terminate the exposure, a backup timer sets a time limit. On installation of a radiographic unit, the dose levels are set, taking into consideration all the components of the imaging chain, i.e. film and screens, imaging plates, film development, preferred tube voltage and filtration, acceptable image noise, etc. This process needs to be carried out for all tube voltages, image receptor and examination types in question. Some products allow for fine manual adjustment to the preset dose level by a density control on the console adapting the density in steps of 10–20%.

Radiographic devices commonly have ionization chambers as AEC detectors positioned immediately in front of the radiographic cassette. The detectors must show no visible radiographic contrast on the image. For low energy X ray units (e.g. mammography, paediatric units), this is difficult to achieve and detectors are therefore positioned behind the image receptor. Solid state detectors are mostly employed in this case.

The position of the detectors is delineated on the table top or wall stand, to assist the operator in patient positioning. As absorption in the patient's body can vary substantially across the beam, the operator can select a detector or a combination of detectors for exposure control, to obtain optimal exposure in the dominant part of the image. As an example, for a chest X ray in posterior–anterior projection, the two lateral detectors positioned under the lung regions are chosen, while in lateral projection, the central detector is selected.

5.4.4. Falling load

To avoid image blurring due to patient motion, short exposure times are mandatory. To produce the shortest possible exposure, the generator starts with the maximum permissible current and, in the course of the exposure, lowers the tube current consistent with tube ratings (falling load). Thus, the tube is operating at the maximum permissible power rating during the entire exposure. In some products, an exposure with falling load can be run at a reduced power setting (e.g. 80% of the maximum power) to lower the stresses. The operator sets the tube voltage, focus size and, if not in AEC mode, the mAs value, but not mA or time.

5.5. X RAY TUBE AND GENERATOR RATINGS

5.5.1. X ray tube

The nominal voltage gives the maximum permissible tube voltage. For most tubes this will be 150 kV for radiography. For fluoroscopy, another nominal voltage might be specified. The nominal focus is a dimensionless figure characterizing the focal size (IEC336). For each nominal focus, a range of tolerated dimensions is given for the width and length of the focus, e.g. a nominal focus of 1.0 allows for a width of 1.0–1.4 mm and a length of 1.4–2.0 mm.

The power rating, P, for a given focus is the maximum permissible tube current for a 0.1 s exposure at a tube voltage of 100 kV. A more practical quantity is the power rating obtained with a thermal preload of the anode (thermal anode reference power) of typically 300 W (see also Fig. 5.11). P depends on the focal spot size and ranges from ~100 kW for 1.5 mm down to ~1 kW for 0.1 mm focus size.

Physical data for the anode include the target angle, anode material and diameter of the disc. The anode drive frequencies determine the rotational speed of the anode. High power loading of the anode requires high rotational speeds. To save the bearings from wear and damage, the speed is reduced for low power settings, as in fluoroscopy. The inherent filtration of the tube is given in equivalents of millimetres Al (see Section 5.6.2).

The heat capacity, Q, of the anode is the heat stored in the anode after arriving at the maximum permissible temperature. Q is equivalent to electrical energy, where $Q = wU_AI_At$, with tube voltage U_A and current I_A, and the exposure time, t. U_A given as a peak voltage is multiplied with a waveform factor, w, to obtain the effective tube voltage (root mean square voltage). W has values of 0.71 for 1- and 2-pulse generators, 0.96 for 6-pulse generators and 0.99 for 12-pulse generators. Q is then given in joules (J). Since early generators were based on single-phase supplies, w was simply set to 1.0 for 1- and 2-pulse generators and 1.35 for 6-pulse generators, giving the heat capacity in another unit, the heat unit (HU), where 1 J = 1.4 HU. The heat capacity of general purpose tubes starts at ~200 kJ, ranging up to >1000 kJ for high performance tubes.

The maximum anode heat dissipation indicates the maximum rate of heat loss typically available at maximum anode temperature. These data depend on temperature and tube type. Tube data also include cooling and heating characteristics (Fig. 5.15).

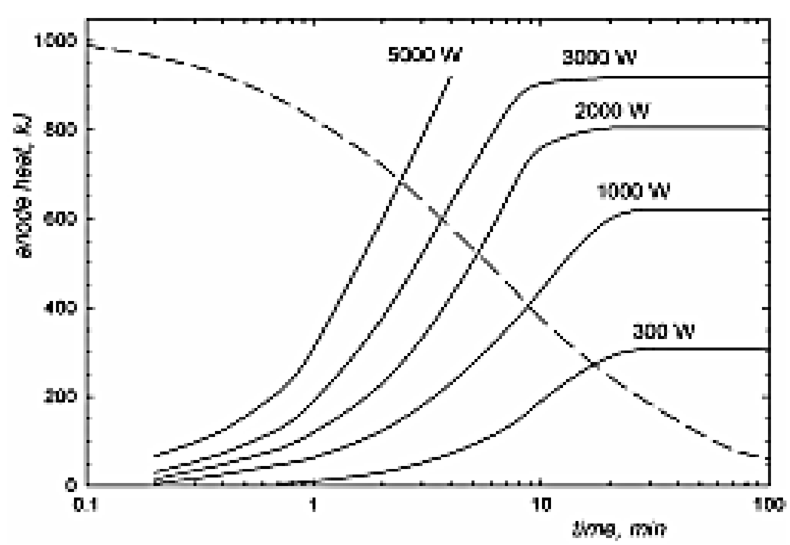

FIG. 5.15. Cooling of the anode (dashed curve) and heat buildup for several constant input

5.5.2. Tube housing

The maximum heat capacity for a tube assembly is typically in the range of 1000–2000 kJ. Maximum continuous heat dissipation describes the steady state of heat flowing in and cooling off. Figure 5.16 shows typical heating and cooling characteristics.

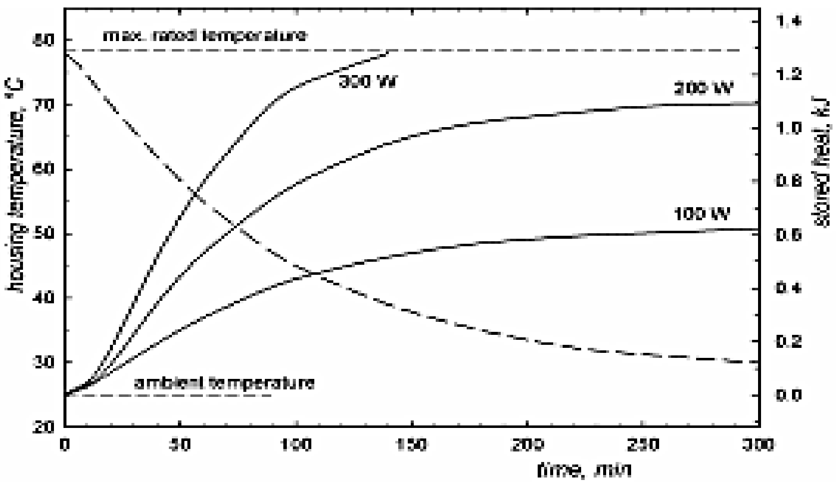

FIG. 5.16. Typical cooling characteristics of a passively cooled tube housing (dashed curve) and heating curves for a constant power input of 100, 200 and 300 W.

The patterns of loading the tube in an examination vary from single radiographic exposures to long high current CT scans, from simple fluoroscopic examinations to long interventional procedures. The tube rating charts contain basic data to estimate the required cooling times. These limits have to be observed, particularly if the control panel gives no indication on actual tube loading or required cooling times. Exposures made by physicists in their measurements can be repeated much more frequently than within the course of a patient examination, and several such high power exposures without observation of the appropriate cooling times can damage the anode and bearings.

5.6. COLLIMATION AND FILTRATION

5.6.1. Collimator and light field

The limitation of the X ray field to the size required for an examination

is accomplished with collimators. The benefits of collimating the beam are

twofold — reduction in patient dose and improvement of image contrast due to a reduction in scattered radiation. A collimator assembly is typically attached to the tube port, defining the field size with adjustable parallel opposed lead diaphragms or blades (Fig. 5.17). To improve the effectiveness of collimation, another set of blades might be installed at some distance from the first blades in the collimator housing. Visualization of the X ray field is achieved by a mirror reflecting the light from a bulb. The bulb position is adjusted so that the reflected light appears to have the same origin as the focal spot of the tube. The light field then 'mimics' the actual X ray field. The congruency of light and X ray field is subject to quality control. One must be aware that some of the penumbra at the edges of the radiation field is due to extra focal radiation.

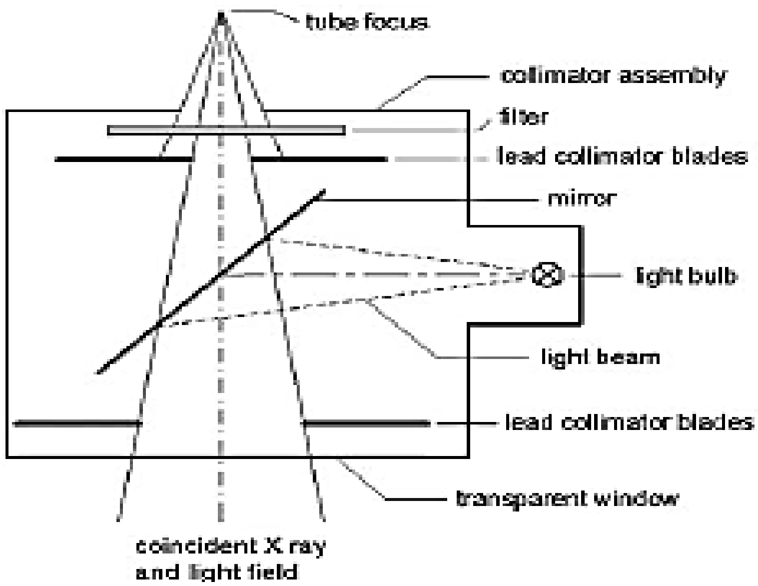

FIG. 5.17. Typical X ray field collimator assembly.

Adjustment of the field size is done manually by the operator, but with a positive beam limitation system, the size of the imaging detector is automatically registered and the field size is adjusted accordingly.

For fluoroscopy, other collimator types are in use, with variable circular and slit diaphragms. In some applications (dental and head examinations), beam restrictors with a fixed field size are typically used.

5.6.2. Inherent filtration

X rays generated in the anode pass various attenuating materials before leaving the tube housing. These materials include the anode, tube envelope exit port (glass or metal), insulating oil and the window of the tube housing. This inherent filtration is measured in aluminium equivalents (unit: mm Al). Aluminium does not perfectly mimic the atomic composition of the attenuating materials present, thus, measurement of the Al equivalent is usually made at 80 kVp (or otherwise the kVp settings should be stated). Typically, the inherent filtration ranges from 0.5 to 1 mm Al. The mirror and the window in the collimator housing also contribute to inherent filtration with an Al equivalent of about 1 mm.

5.6.3. Added filtration

Since filtration effectively reduces the low energy component in the X ray spectrum, a minimum total filtration of at least 2.5 mm Al is required to reduce unnecessary patient dose. Additional filter material is positioned between the tube window and collimation assembly as required. Typical filter materials include aluminium and copper, and in some cases, rare earth filters such as erbium that utilize K edge attenuation effects. Individual filters may be manually selected on some units. In modern fluoroscopy units, filters are inserted automatically, depending on the examination programme chosen.

The effect of added filtration on the X ray output is an increase in the mean photon energy and half value layer (HVL) (see Section 5.7.1) of the beam. As the X rays become more penetrating, less incident dose at the patient entrance is required to obtain the same dose at the image receptor, giving a patient dose reduction. Since image contrast is higher for low energy X rays, the addition of filters reduces image contrast and optimum conditions must be established, depending on the type of examination. Added filtration also increases tube loading, as the tube output is reduced and must be compensated for by an increase in mAs to obtain the image receptor dose required.

In mammography, special provisions concerning filtration are required to obtain the optimum radiation qualities.

5.6.4. Compensation filters

In some examinations, the range of X ray intensities incident upon the image receptor exceeds the capabilities of the detector. Compensation or equalization filters can be used to reduce the high intensities resulting from thinner body parts or regions of low attenuation. Such filters are usually inserted in the collimator assembly or close to the tube port. Examples of compensation filters include wedge filters for lateral projections of the cervical spine, or bowtie filters in CT.