2

X-ray Tube Physics and Technology

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2.1 Working Principle and Types of Medical X-ray Tubes

A comprehensive treatment of the physics, history, technology, reliability, and commercial aspects of diagnostic X-ray sources can be found in Behling (2016). The subsequent chapter will focus on the vacuum technology of X-ray tubes for medical imaging.

2.1.1 A Sample Rotating Anode X-ray Tube

Although the generation of bremsstrahlung (braking radiation) is comparatively inefficient, it has been, and will continue to be, the agent of choice for medical imaging. No other process of X-ray generation is as versatile in practice and as cost effective as bombarding compact targets of high atomic number and density with electrons.

Figure 2.1 is a picture of visible thermal radiation from the thermally excited area next to an X-ray focal spot of a rotating anode tube with a glass frame, which is heated by electron bombardment. The fraction of the intensity of created X-rays divided by the kinetic power of the incoming electron beam, which is in the order of one percent, and the ratio of the space angle that the used beam covers compared with the total isotropic radiation field, about one percent as well, yields an overall efficiency of energy conversion in the region of only $10^{-4}$. A tube for computed tomography (CT) generates as much light intensity as a single consumer LED, but requires input in the order of 100 kW. A total of 99.99% of this energy is wasted as heat. Conrad Roentgen’s first glass tubes already suffered from excessive heating, see Chapter 7. The vast majority of his tubes imploded or otherwise broke; certainly many of them upon thermally overstressing the glass wall which—surprisingly—acted as the initial target.

Unlike Roentgen’s hardly visible glass wall, heat generated in a high power tungsten anode usually shows up well, as Figure 2.1 reveals. The rotating anode shown is facing downward, located about an inch above the cathode at the bottom. In the middle of the image, the conical surface of the anode shows a narrow white glowing rectangular zone, which is about a millimeter wide in the tangential direction and about 10 mm long in the radial direction. It extends radially close to the outer rim of the anode. The center line of the invisible X-ray focal spot where electrons impact the target and the center line of the visible glowing rectangular area are oriented in parallel, but askew by about half a focal spot width. An adjacent comet tail of heat radiation from the focal track tails to the left and further about the center axis of rotation. Overlaid on this picture is a reflection of visible thermal light from the glowing electron emitter coil of the cathode at the bottom of the picture. This reflected light has a color temperature of about 2200°C, the temperature of the filament coil of the cathode. The exact origin of the heat is a radially extending focal spot where the beam of electrons from the cathode impinges on the conical tungsten–rhenium surface of the anode. The electron beam has a nearly rectangular cross-section in the plane of the focal spot. The (invisible) X-ray focal spot, where the X-rays emerge, is slightly wider than the white glowing hot zone where thermal radiation emerges with relatively high intensity. Upon passage of the electron beam from the right to the left, the cool target material heats until its thermal radiation becomes visible. The tail provides information on the speed of the heat spread inside the target. The focal spot visibly cools down during the period of rotation of about 20 milliseconds. Immediate heat diffusion from the very top of the bright electron interaction zone within the focal spot in the normal direction to the surface causes a more rapid fall of the surface temperature of the tungsten target by about 1000 Kelvin right after departure from the electron beam. A centimeter along the focal spot track of 80 mm

![Figure 2.1](image-url)

**Figure 2.1** Thermal light emission from the rotating anode of a glass X-ray tube in operation.
diameter corresponds to about 800 microseconds of motion of an element of the target. The sharp left boundary of the hot zone is indicative of a cooling rate of at least 1000 Kelvin within less than a single dwell time of tungsten under the electron bombardment in the focal spot. The width of this sample spot amounts to about 1.5 mm. The cooling rate is in the order of 10 K/µs in this case. Given the well-known heat conductivity and capacity of tungsten, this indicates that the heat is generated in a very shallow volume close to the surface. Indeed, under practical conditions of tube voltages between 40 and 150 kV, useful X-rays are generated up to about two micrometers deep in the target. The depth of total electron penetration amounts to about 10 µm. This results in a relatively small heat capacity of the heated volume and large heat conduction to deeper material layers. The photo was captured during the initial part of the exhaust procedure with a nearly virgin tube sitting on a vacuum pump (Figure 2.1).

The working principle of an entire X-ray tube may be explained with a sample glass X-ray tube, as depicted in Figure 2.2.

X-rays emerge upon impact of electrons from the cathode and their interaction with the nuclei of the target atoms. Photons fan out from here in all directions with nearly isotropic intensity. Only a small fraction enters the used beam, which is defined by a set of apertures and X-ray filters, see Figure 2.3. X-rays emerging sub-surface have to traverse the anode material. Rays close to the anode shadow are particularly attenuated and spectrally shaped by the tungsten target. The form of the lack of X-ray intensity in the polar diagram of Figure 2.4 gave rise to the term “heel effect.” If the highest energy of photons generated exceeds the k-edge of tungsten (tube voltage greater than 70 kV corresponding to a k-edge at about 70 keV), high energy photons are cancelled by intrinsic filtration. Soft radiation is strongly attenuated in directions close to the anode shadow.

The cathode and anode are placed in an ultra-high vacuum. The cathode usually comprises coiled tungsten wires, which can be directly heated by an auxiliary current. Electrons are thermally excited. Those which overcome the barrier of the work function of the material (in this case, tungsten with about 4.5 eV) are released into the vacuum. Metallic electrodes shape the electrostatic field such that the charge carriers are pulled off the cathode and hit the anode in the specified region of the focal spot. The anode disc is a sintered compound of tungsten with some percentage of rhenium at the top. The bulk material underneath consists of an alloy of molybdenum, zirconium, and titanium. The anode disk is mounted with a thermally isolating (usually hollow) shaft on a bearing system, comprised in this sample case of a ball bearing system in a vacuum. Members of the ball bearing system made of steel are coated with thin layers of a solid material like lead or silver, which prevent cold-welding or pitting. The lifetime of ball bearings in a vacuum is rather limited. The anode therefore has to be stopped when the tube is idle and accelerated before the next exposure. Due to a lack of significant

FIGURE 2.2  Left: cut view of a current rotating anode tube housing assembly for general radiography. Top right: evacuated glass X-ray tube. Used radiation passes through an “X-ray port” or X-ray window in the center.
heat conduction through the point-like contacts of balls and raceways, the exemplary anode solely cools by heat radiation. 

Figure 2.5 shows a more rugged type of rotating anode tube with a metal center section.

Vacuum “inserts” are enclosed by protective housings. Lead or other high-Z material covers the enclosing aluminum housing internally and shields-off unwanted leakage radiation. As discussed, bremsstrahlung from a thick reflection target emerges nearly isotropic in all directions. X-ray lenses are not available for the hard X-rays with spectra useful for human imaging. Only a small amount of the used radiation emerges under an angle, which in practice amounts to between nearly zero and about 30° from the truncated cone of the anode disc. In addition to the target-intrinsic filtration, this radiation is further spectrally shaped by the glass or beryllium wall of the tube insert, a thin layer of insulating oil and a slab of, typically, aluminum. Figures

FIGURE 2.3 Close-up of a cut model of the X-ray port of a metal center section tube with beryllium window.

FIGURE 2.4 Polar diagram of the X-ray intensity distribution for a Philips SRO 2550 rotating anode tube. Electrons impinge at an angle of 15° on a target which has the same anode angle. Triangles: new tube, as processed; tilted squares: aged tube. Beam filtration: 2.5 mm aluminum equivalent plus 20 mm aluminum emulating patient filtering. An extrapolated isotropic distribution is overlaid with the measured data for comparison.
2.5 and 2.6 show exemplary cut models of Varian tube assemblies for CT. For these constructions, glass is replaced by a metal center section. The tube in Figure 2.6 has a single high-voltage terminal for the cathode only, insulated by a ceramics insulator capable of isolating a tube voltage of 140 kV. The anode is on ground potential, a technology that Varian calls anode end grounded. The graphite-backed anode is finned, as was the first ever rotating anode from Philips (see Chapter 7). A direct oil or water cooled scattered electron trap reduces the heat loading of the anode by nearly 40%.

2.1.2 Rotating Frame Tubes

By 2003, Siemens realized an idea created in the late forties by Waterton and Metropolitanvickers, UK, see Waterton (1946), and began commercializing rotating frame tubes for CT, see Schardt et al. (2004) and Figure 2.7. This platform of tubes comes with a rather small anode of only about 120 mm diameter, which rotates as a part of the entire tube frame and is in direct contact with oil. The ball bearings run well lubricated in oil instead of in an “aggressive” vacuum. The tight thermal connection of the anode to the oil prevents the bulk anode temperature from rising above a few hundred degrees centigrade. However, large thermal gradients appear between the focal spot track and the backside interface to the oil.

On the other side of the tube, a circular-symmetric flat electron emitter in the center emits a beam of electrons into the vacuum, which are first accelerated along the tube axis. The X-ray focal spot is kept at a fixed position relative to the tube housing by a stationary magnetic dipole field. The electron trajectories are bent outwardly during passage of the dipole field in the region of the visible bottle-neck and hit the anode about 5 cm off axis.
Additionally, an electrically controllable magnetic quadrupole field shapes the electron beam.

Axial \( z \)-deflection of the focal spot has been introduced with the Straton® tube as a novel, very beneficial option. There is a price to pay, however. Hydrodynamic friction of the fast rotating tube insert requires several kilowatts of driving power, \( P_{\text{rotor-drive}} \), to keep the rotor at a high speed of 200 Hz.

### 2.1.3 Stationary Anode Tubes

Figure 2.8 is a close-up of a stationary anode tube, which comprises a fixed tungsten target of about one millimeter thickness brazed into a copper stem. Compared with the use of ball bearings, stationary or fixed anode tubes provide better heat conduction from the focal spot to the ambient surroundings. The continuous heat dissipation of the approximately 10 cm long tube shown in the figure is nearly equal to the rating of a much more expensive rotating anode tube shown in Figure 2.2. The downside of using a fixed target is a lack of permissible density of instantaneous pulse power input at the target surface. Compared with rotating anode tubes, this figure is about two orders of magnitude smaller for stationary anode tubes. Exposure times may be one or two orders of magnitude longer. Stationary anode tubes serve medical applications of low instantaneous input power paired with moderate continuous power, for example, for surgical C-arms or dental X-rays.

### 2.2 X-ray Source Assembly

According to Food and Drug Administration (USA) terminology, an X-ray source assembly is comprised of a tube (vacuum electronics component) and tube housing (X-ray shield, mechanical protection, terminals), which both make up the tube housing assembly, and a beam limiting device (collimator), as depicted in Figure 2.9.

#### 2.2.1 Protection against Implosion and Explosion

A medical device must be intrinsically safe against hazards caused by malfunctioning of any of the components involved. Anode discs may burst. Glass tubes might implode. Metal center section tubes could even explode after oil leakage and access of air to hot electrodes or electrical discharge. The tube housing protects user and patient against all these potential hazards. Type tests under worst case conditions ensure the integrity of the tube assembly under all thinkable circumstances, based on exhaustive failure mode and effects analysis.

Technically, an onion-like concentric design of a ductile metal center section and ductile tube housing envelope is preferred.
The metal center section catches expelled parts, deforms, and dissipates their kinetic energy.

2.2.2 X-ray Shielding

Lead-lining or enclosure of the X-ray generating focal spot by compounds of a material with a large atomic number and density shields users from unwanted radiation, see for example, Figures 2.2, 2.3, and 2.5. For compliance with US code 21 CFR Part 1020 2005, the housing shield, the X-ray port, and the shielding of the beam limiting device have to be engineered and assembled such that the leakage radiation from the X-ray source assembly would never exceed a local maximum of 0.88 mGy air kerma in one hour at a distance of one meter from the focal spot. Leakage technique factors stated in the accompanying documents specify firstly, and most importantly, the maximal allowed tube voltage of operation and secondly the charge which a tube may conduct during X-ray production within one hour of operation.

2.2.3 Beam Limitation

The definition of the X-ray fan is achieved in two subsequent steps. To exclude most of the off-focal radiation that emerges from the anode entering the used X-ray beam, an upstream aperture is placed in a fixed position proximal to the focal spot. Using this device only, large geometrical magnification would cause penumbra effects and poor definition of the radiated area due to the non-vanishing size of the focal spot. Thus, in addition, the most accurate definition of the radiation field is provided by other X-ray opaque blades at a distance of circa 20 cm from the focal spot inside the beam limiting device, simply—but physically incorrectly—called the collimator. Figure 2.10 illustrates the principle of beam forming. US code 21 CFR Part 1020 2005 requires:

Neither the length nor the width of the X-ray field in the plane of the image receptor shall differ from the corresponding dimensions of the selected portion of the image receptor by more than 3% of the SID (remark: i.e., source-image detector distance) when adjusted for full coverage of the selected portion of the image receptor.

Collimators in whole body CT usually define fields of view of about plus and minus 25–30 degrees in plane and between one and about 16 degrees axial. Their axial collimation determines...
the patient coverage per rotation and is usually dynamically adapting to the rotation phase of the gantry. Beam shaping comprises delimiting and also smoothing the flux of radiation across the X-ray fan. A special attenuator, the often called bow-tie filter, primarily weakens peripheral X-rays. Another device sometimes used to even out the X-ray intensity at the detector is a heel effect filter, which is usually included in the X-ray window of the tube housing.

2.2.4 Beam Quality

Tube housings provide part of the minimal required X-ray filtration that an X-ray source assembly has to deliver. Trade-off between avoiding unwanted soft radiation, guaranteeing exact compliance with regulatory requirements, best X-ray flux, and robustness of the equipment against false assembly, in particular during tube replacement and assembly of tube housing and beam collimator, has led to different solutions. According to International Electrotechnical Commission IEC 60601-1-3, ed. 2.0, the beam quality shall be at least equivalent to 2.5 mm of aluminum in terms of half-value layer. Permanent irremovable filters, for example, the window cup in Figure 2.3, must never undercut 0.5 mm Al equivalent. Fixed added filters shall not render filtration less than 1.5 mm Al. A beam limiting device may add filtration. Therefore, the filtration of the tube assembly alone may undercut the 2.5 mm threshold. However, then it must be ensured that a compatible collimator is mounted. To avoid any uncertainties, some manufacturers like Philips decided to deliver tube assemblies (excluding the collimator) with generally at least a full 2.5 mm Al equivalent. The beam limiting device may then add an aluminum equivalent of a few tenths of a millimeter and attenuate the beam further. The X-ray flux is somewhat reduced for the sake of patient safety. Other manufacturers impose more strict requirements on the compatibility of the tube housing assembly and the attached collimator. Service staff are then in charge in the clinic.

2.3 Cooling

Table 2.1 illustrates various sources of heat in a typical X-ray source. Direct air convection, as depicted in Figure 2.2, has still been by far the most popular way of cooling. Rubber bellows allow for oil expansion. Internal convection in oil translates to external convection in air. Gravitational circulation of oil is essential. According to the standard (IEC 60601-2-28 2010) the

\[ \text{Temperature of the painted surface of an X-RAY TUBE ASSEMBLY which can unintentionally be touched during intended use \ldots shall not exceed 85}^\circ\text{C}. \]

Systems in mixed use with a large portion of fluoroscopic applications benefit from doubling the heat dissipation capability of pure air convection with small water coolers. Heat exchanging surfaces inside the tube housing assembly are cooled by an external water supply. Other than general radiography, interventional angiography and CT application result in a required average heat dissipation in the order of up to about six kilowatts, see Table 2.1, which can only be handled by enforced oil cooling, see Figure 2.11 and Table 2.2.

2.4 Tube Components in Detail

2.4.1 The Cathode

The rotating anode shown in Figure 2.12 was heated by electron bombardment under high voltage from the cathode below. The snapshot was taken with the tungsten electron emitter coil still at elevated temperature. In an attempt to better visualize the situation without exceeding the dynamics of the camera, the direct heating current through the coil (about six amperes) had been switched off 100 milliseconds before the picture was taken. Figure 2.13 depicts a frontal close-up of a typical cathode head. This sample comprises two distinct emitter coil structures which are embedded in a geometric electrostatic field shaping and focusing electrode, the cathode head. Each coil is capable of producing an X-ray focal spot of a defined size on the anode (small and large focal spot in this case). The cooling time of each of the

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**TABLE 2.1**

<table>
<thead>
<tr>
<th>Source</th>
<th>Typical Maximal Long Term Average Thermal Power Input</th>
</tr>
</thead>
<tbody>
<tr>
<td>X-ray generation</td>
<td>200 W (general radiography)</td>
</tr>
<tr>
<td></td>
<td>1500 W (interventional radiography)</td>
</tr>
<tr>
<td></td>
<td>6000 W (computed tomography)</td>
</tr>
<tr>
<td>Filament heating</td>
<td>80 W</td>
</tr>
<tr>
<td>Rotor drive</td>
<td>100 W (start-stop, ball bearing)</td>
</tr>
<tr>
<td>Internal pumps</td>
<td>700 W (continuous rotation, spiral groove bearings)</td>
</tr>
<tr>
<td>Auxiliary devices (grid switch, other electronics)</td>
<td>50 W</td>
</tr>
</tbody>
</table>

250 μm thick coiled wires amounts to about a third of a second. As thermionic electron emission strongly depends on the emitter temperature (see below), the tube current can be controlled by steering the heating current. Speeding up the rise of the current is typically achieved through over-boosting the heating current (e.g., by imprinting 10 or more amperes for a split second). However, ramp-down can hardly be accelerated. This time is governed by the speed of heat dissipation from the tungsten wire to the ambient surroundings, primarily determined by heat radiation in the case of large tube currents (e.g., several hundred milliamperes) and a high emitter temperature in the range beyond 2200°C. When small tube currents are drawn, for example, through stationary anode tubes of the type depicted in Figure 2.8 and at moderate emitter temperatures below 2000°C, most of the emitter cooling will be provided by heat conduction through the posts which support the wires. Overall, with this widely used technique, significant tube current modulation can be achieved with a frequency of about three cycles per second; the greater is the speed and amplitude of the tube current modulation. Pulsing an X-ray beam with shorter pulses, for example, in the millisecond range, as used for angiography, thus requires other methods.

For historic reasons, the tungsten emitter wire is often called the “filament.” We recall that W. D. Coolidge replaced carbon filaments in light bulbs with tungsten wires. Accordingly, the heating current is usually termed the filament current. Behling (2016) provides more details on this matter.

### 2.4.2 The Work Function of Tungsten

Electrons usually remain inside the emitter metal unless they are thermally excited. A small fraction may then be able to overcome the energetic barrier to the vacuum: the work function. According to the Boltzmann distribution, the probability of finding an electron in an energy state beyond the work function is proportional to

**Table 2.2: Performance of Cooling Concepts**

<table>
<thead>
<tr>
<th>Cooling Concept</th>
<th>Typical Continuous Heat Dissipation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air convection</td>
<td>200 W … 300 W</td>
</tr>
<tr>
<td>Enforced air convection (fans)</td>
<td>350 W … 450 W</td>
</tr>
<tr>
<td>Remote heat exchanger</td>
<td>500 W … 3500 W</td>
</tr>
<tr>
<td>Attached heat exchanger (see Figure 2.11)</td>
<td>2500 W … 6000 W</td>
</tr>
</tbody>
</table>

*Source: See Behling, R. 2016. Modern Diagnostic X-ray Sources. Technology, Manufacturing, Reliability. CRC Press, Taylor and Francis, Boca Raton, FL, USA.*
to $e^{-E/(k_BT)}$, where $E$ is the kinetic energy of the electron relative to the Fermi-level, $k_B$ is Boltzmann’s constant and $T$ is the temperature of the metal. The Fermi-level is defined as the energy of the highest populated state in the absence of thermal excitation. When electrons enter a vacuum, a mirror charge and an attractive electric force appear and contribute to the work function.

Figure 2.14 explains emission physics in more detail. Without considering space charge effects, the Richardson–Dushman equation referenced in this figure fairly well quantifies the tube current as a function of the emitter temperature. This assumption on space charge holds, in particular, for stationary anode tubes which operate anode limited. It is also valid for conventional rotating anode tubes with cathodes, shown above, when the tube voltage is larger than the so called isowatt point. This isowatt point, a voltage, is a measure of the cathode performance in relation to the anode performance. Figure 2.15 depicts a typical emission chart of a sample tube for angiography. The abscissa indicates the filament heating current and the corresponding emitter temperature. The left ordinate shows the tube current depending on the parameter tube voltage. In this typical graph, which represents most tube specifications, the exposure time is set to a standardized nominal value of 100 milliseconds. The right ordinate indicates the heating voltage required to imprint a desired heating current into the cathode terminals of the X-ray tube. In total, the emitter wire, support posts, connecting leads, feed-throughs, and also the high voltage of the leads result in an ohmic relationship between heating voltage and current. Usually, the dominant contributor is the tungsten wire. As is usual for metals, its resistance rises significantly with the emitter temperature, as indicated in the graph. As the heating power is proportional to the wire resistance, and as powering up should be as quick as possible, the cathode is pre-heated in practice, even when the tube is kept idle. The wire temperature is held under about 1800°C by the generator driving electronics such that the rate of tungsten evaporation is low enough to sustain an entire tube life and electron emission will be at the lowest specified level when the high-voltage is switched on.

Figure 2.15 also indicates space charge limited emission for a range of technique factors. In the case of a high emitter temperature and a tube voltage below the isowatt point (about 75 kV in this example), once escaped, the vast majority of electrons are returning from the vacuum back to their origin in the metal. The Child–Langmuir law (Equation 2.1) describes the asymptotic situation where the emission is totally space charge limited:

$$j_e = \frac{4\pi e_0}{9} \sqrt{\frac{2e}{m}} V_t^2 \frac{3}{d^2} \quad (2.1)$$

Charged with the tube voltage $V_t$ between cathode and anode over the distance $d$, electric force pulls the electrons off the emitter. $j_e$ denotes the current density at the emitter surface, $e_0$ the vacuum dielectric constant, $e$ the absolute value of the electron charge, $m$ its rest mass. In this extreme case, neither heating the
tungsten further nor using a different emitter material with a smaller work function would help to increase the tube current. But usually, as shown in the emission chart (Figure 2.15), the Child–Langmuir law “competes” with the Richardson–Dushman law.

In this mixed mode, indicated by the blue squares in Figure 2.15 at the maximum permitted emitter temperature, which the manufacturer specifies for the sake of durability of the tube, coil emitters exhibit an approximately linear decrease of the maximal tube current $I_t$ with the tube voltage. Raising the heating current by just 10% (which means raising the heating power by ca. 20%) doubles not only the rate of evaporation of electrons, when space charge effects are ignored, but also intensifies the evaporation of tungsten atoms by an order of magnitude. As discussed before, the local heating power rises with resistivity. Once a local waist has started to develop, the temperature rises at this point and accelerates the rate of evaporation of material until such a hot spot melts and the filament has “burned out.”

Evaporating electrons extract energy from the filament. The electronic cooling power is proportional to the product of the work function and tube current. Given the large work function of tungsten of 4.5 eV, the energy taken away by the emitted electrons with currents of hundreds of milliamperes may cool the wire with several watt. Although this power loss is minor in relation to the tens of watts for heating, the high sensitivity of the current density with respect to temperature in the Richardson–Dushman regime of saturation emission results in a well noticeable drop of emission of typically 5–10% after switching on the high voltage. The temporal behavior resembles the aforementioned speed of emitter cooling with characteristic times of about a third of a second. The generator heating control has to compensate for this and feed more heating current to keep the tube current stable.

Most of the systems for CT vary the heating current during a scan and modulate the tube current to at least partly balance the varying transparency of the human body under different perspectives. More details on tube current modulation and grid switching can be found in Behling (2016).

2.4.3 Alternatives to Tungsten Thermionic Emitters

2.4.3.1 Reduction of the Work Function

As, according to the Richardson–Dushman law, the tube current depends extremely on the work function of the emitter material, several methods have been devised to reduce it. As an example, a monolayer of thorium on top of the tungsten matrix reduces the work function by about 40%. Thorium diffusion replenishes loss from evaporation and sputtering effects. Usually, tungsten wires are carbonized to control the diffusion speed. Unfortunately, all thorium isotopes are radioactive. Reduction of the work function to 1.16 eV and current densities of up to 400 A cm$^{-2}$ have been reported for dispenser cathodes, see Gärtner (2000), Gärtner and Barratt 2004, Engelsen 2005. The top layer is covered with barium and barium oxide dipoles in a complex with tungsten, scandium, and oxygen. More details can be found in Behling (2016), including a discussion of the challenges to introduce...
those solutions in rugged and not actively evacuated medical X-ray tubes.

2.4.3.2 Carbon Nanotube (CNT) and Graphene Emitters

Multiple source X-ray tubes, comprising electron field emission cathodes, have been presented and tested, for example, in the University of North Carolina, Chapel Hill, for motionless mammography tomosynthesis imaging, see Gidcumb et al. (2014). Instead of overcoming the work function barrier by thermal excitation, electrons are field-emitted from bunches of single-walled or multi-walled carbon nanotubes (CNTs) which allow for pulling currents of a few microamperes each. Quantum mechanical tunneling of electrons through a shortened energy barrier causes current densities at the individual emitters to nearly follow the Fowler–Nordheim law and are about proportional to \((1/Vt)^2 \cdot \exp(-\text{const}/Vt)\), with \(Vt\) being the tube voltage. The barrier is narrowed by applying a high electric field, which is intensified at the apex of each field emitter by geometrical enhancement. In order to achieve a minimum focal spot size and to bear sufficient current, multiple CNTs are arranged to cover cathode areas of several millimeters in length and width. Upon application of an electric pull field, each emitting area generates bunches of electrons which produce projected focal spot sizes of the desired dimensions. The high mechanical strength of CNTs, their exceptional electrical and thermal conductivity, and relatively low sputter rates promise CNTs to be excellent candidates for field emitters. Challenge arises from the necessity to apply pull voltages in the range between 1 kV and 1.2 kV, depending on the individual electron emissivity of the CNTs. A basic vacuum pressure of \(10^{-6}\) Pa inside the tube is recommended, which is achieved by ion getter pumps. Although common for stationary anode tubes, this required maximum gas pressure is about three orders of magnitude below the typical maxima in rotating anode tubes and renders the technology very difficult for these. Field emission-based free electron generation based on CNTs has its primary benefits for use in tubes with stationary targets. More details can be found in Behling (2016). Despite of more than two decades of intense scientific work and remarkable achievements, CNT’s and graphene edge emitters have not yet entered commercial tubes for broad scale medical imaging.

2.4.3.3 Electron Beam Focusing and Metric for the Focal Spot Size

As aforementioned, most conventional tubes equipped with thermionic cathodes offer focal spots of different length and width. Electrons are generated by independent emitter coils. Electrostatic focusing is the simplest and most widely used technique. When these cathodes operate in the saturation emission regime where space charge can be ignored, they produce focal spots of sizes which are invariant of the tube voltage applied. Other than for magnetic focusing, no dedicated control electronics is needed to adapt the magnetic field to the kinetic energy of the electrons. We recall that conventional diagnostic imaging stretches across X-ray spectra generated with tube voltages of between 40 and 150 kV. Even in CT, the range of applied tube voltages has recently gone up to between 70 and 150 kV. Complexity rises when bias voltages are applied to electronically reduce the focal spot width, see Figure 2.21.

**Figure 2.16** is the result of an electronic ray tracing simulation of such an electrostatic focusing cathode, which generates a pair of super-imposed focal spots.

The situation becomes more complicated when space charge is to be considered, as in most rotating anode tubes. **Figure 2.17** depicts focal spots of different sizes, generated by the same cathode and powered by the same emitter coil. It is clearly visible that the width of the focal spot is dependent on the technique factors used. There is no general rule which would describe this type of “blooming.” It will depend on the electron optical design of the focusing geometry. Developers play with various degrees of freedom to optimize the cathode for the desired purpose. The sample electron optics that delivered the focal spots shown in **Figure 2.17** comprise cross-over of peripheral electron trajectories when space charge is minor, as seen in the right focal spot exposure. The picture in the middle characterizes the spot size for the dominating conditions of operation. Space charge partially compensates for the cross-over. The focal spot width is minimal in this mixed mode. The left picture shows blooming caused by large amounts of space charge. Notably, the length is hardly affected due to the projection factor between the physical and optical focal spot, which reduces the impact of edge effects. **Figure 2.18** schematically depicts the camera used to measure focal spot intensity distributions and derive focal spot sizes.

As X-ray lenses are not available for the relevant photon energies, focal spot cameras are based on the camera obscura.
principle. Pinhole or slit apertures define the beam geometry and map the projected focal spot from the anode to a film or a digital pixelated detector. Figure 2.19 is a close-up of the aperture side of such a camera. The pinhole aperture is visible in the middle.

The relevant standard, see IEC 60336 (2005), defines the standard technique factor for compliance, see Table 2.3, which may be cited in this context for convenience and illustration.

The size of the exemplary focal spot shown in Figure 2.17 is specified by the dimensionless number 0.6 on a target with 7° anode angle. Any haze around the core of the intensity distribution is excluded as long as it does not exceed 15% of the maximal intensity.

The dimensions of focal spot width, FS width, and focal spot length, FS length, measured in the axial direction of the tube, that is, in parallel to the axis of rotation of a rotating anode tube, are derived by measuring a pair of line spread functions, \( \text{lsf}_{\text{width}} \) and \( \text{lsf}_{\text{length}} \). \( \text{lsf}_{\text{width}} \) is taken with a slit camera aligned orthogonal to the width direction and \( \text{lsf}_{\text{length}} \) is taken with a narrow slit aligned perpendicular to the length direction. An IEC compliant slit camera, similar to the one shown in Figure 2.18, comprises an X-ray opaque slab with a 10 µm wide slit aperture positioned between the focal spot and an image receptor in a pre-defined position. This produces an enlarged image on the pixelated detector. For the statement of compliance, pre-defined enlargements of between one and at least three are mandatory, depending on the size of the spot. The line spread functions are generated by scanning raw data from an image receptor, usually by electronically integrating a two-dimensional detector readout perpendicular to the scanning axis and creating a one-dimensional distribution by summation over the orthogonal direction. The enlargement of the camera above is considered such that the resulting output line spread functions appear as if taken in the focal spot plane without magnification. The spatial distances between the 15% values of the corresponding line spread functions relative to their maxima are stated as focal spot width, FS width, and focal spot length, FS length. Table 2.4 cites the maximum values of the focal spot dimensions as defined by the IEC standard 60336, ed. 4. Figure 2.20 illustrates an exemplary measurement of a focal spot size and a point spread function, that is, the X-ray intensity profile of the focal spot.

FIGURE 2.17  Focal spot blooming. Low tube voltage and rather low tube current (60 kV, 100 mA) results in the appearance of a haze which widens the focal spot. The focal shot shrinks at the desired point of operation of this exemplary tube (middle focal spot image). Space charge and pre-defined (charge-less) cross-over design of trajectories are in balance. Space charge effects nearly disappear at higher tube voltage and reduced current. Cross-over becomes visible.

FIGURE 2.19  Sample focal spot camera with a 30 µm pinhole, see Figure 2.18.
TABLE 2.3
Loading Factors for Focal Spot Measurement and Statement of Compliance

<table>
<thead>
<tr>
<th>Nominal X-ray Tube Voltage</th>
<th>Required X-ray Tube Voltage</th>
<th>Required X-ray Tube Power</th>
</tr>
</thead>
<tbody>
<tr>
<td>Radiography other than computed tomography</td>
<td></td>
<td></td>
</tr>
<tr>
<td>U &lt; 75 kV</td>
<td>Nominal X-ray tube voltage</td>
<td>50% of the nominal X-ray tube voltage as specified by IEC 60613</td>
</tr>
<tr>
<td>75 kV &lt; U &lt; 150 kV</td>
<td>75 kV</td>
<td></td>
</tr>
<tr>
<td>150 kV &lt; U &lt; 200 kV</td>
<td>50% of the nominal X-ray tube voltage</td>
<td></td>
</tr>
<tr>
<td>Computed tomography</td>
<td>120 kV</td>
<td></td>
</tr>
</tbody>
</table>


The IEC values are nominal descriptors. They never bear units. This is often confused in many of the discussions on spatial resolution impacted by X-ray tubes. When a single focal spot value is stated (which comprises information of focal spot width and length at the same time), the length dimension is specified by a factor of ca. 1/0.7 larger than the width for focal spots >0.25. One reason is filament cooling by heat dissipation through the support posts of the emitter coil, which usually results in a smooth decline of the X-ray flux from the edges of the focal spot in the length direction. For CT and increasingly also for tubes for intervention radiography this may be disadvantageous in view of the thermal loading of the focal spot track and the achievable total photon flux. Instead, a square-like shape is preferred. Electrons emerging from the periphery of the emitter coils are deflected by end tabs towards the center of the focal spot. For these cases, where length and width are to be stated independently, the IEC standard offers to state length and width separately in the format "<Nominal width> × <Nominal length>“; and the maximal length of the focal spot is read from the width-column of Table 2.4.

The absolute of the Fourier transform of the line spread functions is the basis for stating the modulation transfer function (MTF) of an X-ray tube. MTF has one interpretation as the degree of modulation transfer as a function of space frequencies and another as the limit of the special resolution. This single number is the space frequency read at the first undercut of the 10% (of maximum) levels of the modulation function. Both standard directions, length and width, are first considered separately. The maximum of both is then stated as the single number MTF value of the tube, using the technique factors defined by the IEC standard. A more detailed treatment of scaling the MTF considering the system magnification delivers Behling (2016), see also IEC 60336 (2005).

### TABLE 2.4
Maximum Permissible Dimensions for Nominal Focal Spot Values

<table>
<thead>
<tr>
<th>Focal Spot Dimensions, Maximum Permissible Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>Nominal Focal Spot Value (Dimensionless)</td>
</tr>
<tr>
<td></td>
</tr>
<tr>
<td>0.1</td>
</tr>
<tr>
<td>0.15</td>
</tr>
<tr>
<td>0.2</td>
</tr>
<tr>
<td>0.25</td>
</tr>
<tr>
<td>0.3</td>
</tr>
<tr>
<td>0.4</td>
</tr>
<tr>
<td>0.5</td>
</tr>
<tr>
<td>0.6</td>
</tr>
<tr>
<td>0.7</td>
</tr>
<tr>
<td>0.8</td>
</tr>
<tr>
<td>0.9</td>
</tr>
<tr>
<td>1.0</td>
</tr>
<tr>
<td>1.1</td>
</tr>
<tr>
<td>1.2</td>
</tr>
<tr>
<td>1.3</td>
</tr>
<tr>
<td>1.4</td>
</tr>
<tr>
<td>1.5</td>
</tr>
<tr>
<td>1.6</td>
</tr>
<tr>
<td>1.7</td>
</tr>
<tr>
<td>1.8</td>
</tr>
<tr>
<td>1.9</td>
</tr>
<tr>
<td>2.0</td>
</tr>
<tr>
<td>2.2</td>
</tr>
<tr>
<td>2.4</td>
</tr>
<tr>
<td>2.6</td>
</tr>
<tr>
<td>2.8</td>
</tr>
<tr>
<td>3.0</td>
</tr>
</tbody>
</table>


#### 2.4.3.4 Off-Focal Radiation

Unfortunately, unwanted radiation may originate from a relatively large area around the zone of primary electron impact. It may wash-out image contrast or even mislead diagnostics by bleeding artifacts. One cause of this radiation is the amount of backscattered electrons from the anode after first impact. They may be mirrored back to the anode and still carry a significant kinetic energy. This effect will be discussed below in the context of the thermal consequences, see Figure 2.29. Another factor is Compton scattering from material in the X-ray window area. In particular, elements of low atomic number, like beryllium and aluminum, and oil may cause significant scatter of X-rays from the primary beam. Scattered radiation from the tube window may add to the primary radiation of the focal spot. If not efficiently shielded off by apertures, minimized by using thin sheets of material only, capturing backscattered electrons, off-focal radiation may amount to about 10% of the primary. An aperture which helps narrow the beam and prevent off focal radiation from reaching the image receptor is placed proximal to the focal spot, as can be seen in Figure 2.2. This aperture primarily helps...
against radiation created upon second impact on the anode of backscattered electrons and is a valuable means in glass tubes. It is less efficient against Compton scatter from components of the X-ray window.

### 2.4.3.5 Special Cathode Features

A cathode, which allows narrowing of the active emitting area and of deflection of the center of the focal spot on the anode, is shown in Figure 2.21. The single emitter is enclosed by isolated control electrodes. Narrowing the electron beam is achieved through negative biasing with respect to the emitter coil. Asymmetric biasing results in deflection in the focal spot width direction, as demonstrated in Figure 2.22. When the biasing voltage exceeds a threshold, usually in the order of three kilovolts, the electron emission will be totally suppressed. Tubes for angiography typically operate many meters distant from the high-voltage generator. High-voltage cables may be as long as 40 meters, and cable capacitance in the order of several nanofarad. In cardiology, heart motion is “frozen” using X-ray pulse lengths of down to three milliseconds. An issue of extra patient dose arises from this concept, in particular in pediatric diagnostics where tube currents are kept intentionally small and pulses short. Cable capacitance is discharged through the tube only, which may last several milliseconds. Long transition times with an improper X-ray spectrum may result. Grid switching is an excellent means to avoid these undesired X-ray spectra.

A sophisticated cathode of a General Electric (GE) HD tube for CT is shown in Figure 2.23. An additional electrode on both sides of a D-shaped tungsten coil emitter allows for focusing and modulating the electron emission. Operation under high and low tube voltage, as it occurs for dual energy CT by fast modulation of the tube voltage between 80 and 140 kV, would suffer from space charge limitation of the electron emission at low tube voltage or overloading the focal track at high tube voltage if there were no means to steer the emission as quick as the tube voltage, that is, within a time frame in the order of a few hundred microseconds or below. GE’s flattening of the surface of the coil improves the emittance and the emission at low voltage. Thermal control of the emitter temperature, as otherwise used for dose

![FIGURE 2.20](image)

**Definition of the focal spot dimensions, according to the standard IEC 60336 2005.** Left and center: a pair of line spread functions, lsf$_{width}$ and lsf$_{length}$, is measured with a slit camera. In two measurement cycles, the slit is aligned orthogonal to the width direction and to the length (axial tube) direction. Raw data are scanned from an image receptor by electronically integrating the two-dimensional detector readout perpendicular to the scanning axis and forming a one-dimensional integral distribution. Right: a schematic pinhole exposure is indicated to the right, which represents the point spread function, the spatial resolved electron current density in the focal spot, see Figures 2.17 and 2.18.

![FIGURE 2.21](image)

**Picture of the single emitter cathode of a tube for computed tomography.** Bottom: control electrodes left and right of the tungsten coil electron emitter for deflection of the generated electron beam as well as focusing it by electrode charging with respect to the emitter. Top: electron ray tracing simulation for this cathode.
control, as aforementioned, would be far too slow by two orders of magnitude.

2.4.3.6 Advanced Electron Optics

Since the turn of the century, novel focusing technology for medical X-ray tubes has become available on a broad commercial scale, which has significantly reduced space charge limitations. Still based on thermionic emission, flat directly heated tungsten sheet emitters were combined with magnetic focusing and deflection of the electron beam. The flat electron emitter had already been in use by Siemens since the nineties for mammography tubes and for the angiography tubes MegalixCat Plus® and later Gigalix®. It was then adapted for the rotating frame CT tube Straton®, sold from 2003 onwards, see below. Philips enlarged the size of the emitting surface to 0.5 cm². This is about half of an order of magnitude more than the area used with coil emitters of the type seen in Figure 2.13. The company added a magnetic double quadrupole focusing system, as depicted in Figure 2.24. Unlike Siemens’ Straton® tube, which employs a single quadrupole system, the double quadrupole unit of the Philips iMRC® tube, see Figure 2.25, launched in 2007, allows for an unprecedented high electron trajectory compression factor. Siemens adopted this technology in their latest Vectron® tube for the Somatom Force CT system, launched in 2013.

First, electrons released from the flat emitter are accelerated while the cross-section of the beam hardly changes. Then, the beam is widened by the left magnetic quadrupole and finally compressed by the right one in Figure 2.24. The figures also explain the important capability of z-deflection and x- (width) deflection by two extra dipole coils. The resulting emission characteristics underscore the success of the effort. Space charge limitation at high emission currents and low tube voltages has completely disappeared, as well as blooming effects. The magnets are actively controlled, as the magnetic field strengths have to be adapted to the tube voltage. Thus, controlling this kind of sophisticated electron optics requires intelligent high-voltage generators.

Philips has also introduced a hybrid solution for axial z-deflection for mid-tier CT systems. Electrostatic focusing and deflection perpendicular to the tube axis in the width (x-) direction, as shown for the tube in Figure 2.26a, are combined with radial magnetic deflection along the direction of the tube axis in the z-direction, as shown for the tube in Figure 2.26b. The latter tube frame is formed as a narrow bottleneck between cathode and anode, encompassed by the yokes of a magnetic dipole.
Varian also unveiled a tube capable of magnetic “z-toggling” and fast modulation of the beam intensity, the MCS 7500 CT tube. Details about transition times and limitations of the tube power associated with x-deflection are treated in Behling (2016).

### 2.4.3.7 Backscattered Electrons

Under medical imaging conditions, and even at normal impact, about half of the incoming primary electrons are backscattered from a tungsten target, as shown in Figure 2.27. This is caused by the rapid angular diffusion that electrons experience during scatter at highly charged atomic nuclei close to the metal–vacuum interface. Oblique impact yields even higher rates. In order not to lose those electrons from X-ray generation, the impact angle should be about normal. Some tube types violate this rule and may suffer from inferior conversion factors. A significant share of backscattered electrons have a kinetic energy close to the primary, as Figure 2.28 demonstrates. The fastest may generate significant intensity of bremsstrahlung upon second impact, causing off-focal radiation, as aforementioned. Figure 2.29a shows the tangential and radial deflection of the focal spot in a CT tube as a hybrid of electrostatic and magnetic deflection.

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**FIGURE 2.25** Single polar Philips iMRC® tube for the Brilliance iCT® and the spectral detection IQon® scanner families. (Image courtesy of Philips.)

**FIGURE 2.26** Tangential and radial deflection of the focal spot in a CT tube as a hybrid of electrostatic and magnetic deflection. (a) Configuration with electrodes parallel to the emitter coil providing tangential (width, x-) deflection. (b) Tube with a bottleneck and a pair of magnetic yokes for magnetic radial (length, z-) deflection of the electron beam in addition to the electrostatic tangential (width) deflection.
The bi-polar Philips metal ceramics SRC® tube series from 1979 comprised a molybdenum aperture as electron capturing device, see Figure 2.30. This helps capture at least the most energetic electrons in this bi-polar solution. In most metal center section tubes, the area around the X-ray window and the window itself is subject to heavy bombardment of backscattered electrons. Cooling fins, as depicted in Figure 2.31, are then often used to bring the temperature down below the threshold of window carbonizing by cracking the cooling oil.

The best electron optic solution against second impact on the tungsten target is shown in Figure 2.29b, realized, for example, in the Philips iMRC® tube, see Figure 2.25. Less effective solutions have been commercialized earlier by Varian in the late nineties, see Figures 2.6 and 2.32, and also by GE, see Figure 2.33. In these tubes, a substantial portion of the many electrons, see Figure 2.27, which are backscattered from the focal spot, are forced to return to the anode. These electrons experience the negatively charged cathode. Instead, the grounded anode of the Philips iMRC® tube is positioned in a nearly field-free space behind a grounded electron drift tunnel. Given the electron energy distribution shown in Figure 2.28, minor space charge in the electron beam with a potential in the order of a maximum of a hundred electron volts can be ignored. No significant repelling electric field hinders the ballistic flight of backscattered electrons away from the vicinity of the focal spot, which ends on the surface of an electron trap. This device is on one hand made from low- to moderate Z material like molybdenum; secondly, it geometrically shields X-ray photons, which are created during electron impact on its surface from entering the used beam. Less than a percent of off-focal radiation from multiple electron scattering is generated. About the same amount of off-focal radiation may be caused by photon scattering in the material of the X-ray window and the CT system. Intensity and spectral characteristics of this unwanted radiation are a function of the atomic numbers and the thicknesses of the materials in the X-ray fan beam.

2.4.4 The Anode

The anode primarily determines the tube performance. This text will focus on rotating anodes. Behling (2016) compares stationary and rotating targets in more detail. Compared with heat diffusion, realized with stationary anodes, heat convection, which means feeding cold material into the focal spot, allows for about two orders of magnitude higher densities of input power. Following the first commercialization of finned rotating anodes by Bouwers at Philips in 1929, the quest for increasing instantaneous photon flux has triggered heavy investment in rotating anodes and their cooling, in particular after the invention of CT. At present, target diameters for medical imaging tubes range between 60 and 238 mm, and rotational speeds between about 50 and 200 Hz.

Figure 2.34 presents a selection of anodes for CT tubes with their key characteristics. Although only the anode disks are...
shown, by definition, in this context of the IEC standard, “anode” means the entire unit which is charged with anode potential, which is either positive, as in the original language, or ground potential. Thus, for anode grounded solutions, the definition is somewhat vague when it comes to specifying figures like heat content. Data characterize the anode disk and heat conduction to the ambient through the bearing system, respectively, directly into the oil (in the case of the rotating frame tube). Figure 2.35 is a close-up of a modern segmented high-performance graphite backed metal anode disk for the same purpose, which illustrates the multitude of aspects a modern anode disk has to comply with.

### 2.4.4.1 Thermal Balance

Tube development starts with simulating the thermodynamics of the anode system under the assumption of the targeted application. Whilst ensuring the mechanical integrity of a tube requires numerical methods of temperature simulation for the most part, an approximate analytical treatment is available for the focal spot.

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**FIGURE 2.30** First all-metal–ceramics rotating anode tube with electron capturing aperture, Philips SRC 120 0612 from 1979. Energy of backscattered electrons is kept away from the anode by a molybdenum aperture as part of the tube frame, which is at ground potential. (Image courtesy of Philips.)

**FIGURE 2.31** Finned beryllium window of a metal center section tube for computed tomography.

**FIGURE 2.32** Cut view of the casted tube frame of the anode end grounded Varian MCS® tube, see Figure 2.6, with direct oil-cooled scattered electron trap (beryllium X-ray window removed before cutting).
2.4.4.2 Focal Spot Temperature

Müller (1927) was the first to deliver an analytical formula for the temperature rise in the focal spot of a rotating anode tube. He assumed pure surface heating of a semi-infinite target by a Gaussian electron beam current distribution. Bowers and Oosterkamp of Philips treated more practical cases of rectangular focal spots of short pulse duration on stationary targets, see Oosterkamp (1948a), rotating targets, see Oosterkamp (1948b), and by applying continuous loads, see Oosterkamp (1948c). Several corrections have been proposed since these early papers, including consideration of voltage-dependent volume heating and results from Monte Carlo simulations. Details can be found in Behling (2016). However, the simpler analytical approximations are still widely used.

For a more detailed discussion of the thermodynamics, a labeled thermal image of a rotating anode in operation is shown in Figure 2.36. As a consequence of approximately two-dimensional heat diffusion, the peak focal spot temperature on a rotating anode rises proportionally to the square root of the dwell time, $\Delta t$, of an element of the focal track under the electron beam, see Oosterkamp (1948a). $f_{\text{anode}}$ is the rotational frequency of the anode, $d_{\text{track}}$ the diameter of the focal track, $FS_{\text{width}}$ the focal spot width, $FS_{\text{length}}$ its projected length, $\alpha$ the anode angle, $T_{FS}$ the maximal surface temperature in the focal spot, $\Delta T_{FS}$ the temperature rise that a small element of tungsten experiences upon passage of the electron beam, $T_{\text{track}}$ the temperature of an element of the focal track just before entering the electron beam. The Müller–Oosterkamp law then states, using $T_{FS} = T_{\text{track}} + \Delta T_{FS}$ for the temperature rise in focal spot,

$$\Delta T_{FS} = \frac{2 \cdot P}{\pi \cdot FS_{\text{length}} \cdot \sin(\alpha) \sqrt{\lambda \rho c_p \cdot FS_{\text{width}} \cdot f_{\text{anode}} \cdot d_{\text{track}}}}$$  \hspace{1cm} (2.2)

$P$ denotes the net power which the electron beam imprints in the focal spot, after deduction of the power taken out by back-scattered electrons, $\lambda$ the heat conduction, $\rho$ the target density, $c_p$ its specific heat (per mass unit). Figure 2.37 illustrates, in an exemplary way, the repetitive temperature increments in the focal track per revolution with 12 periods of anode rotation.

In addition to analytic solutions for the focal spot temperature, numerical finite element analysis is often used to calculate the depth profile of the temperature distribution and the induced thermomechanical stress, see Figure 2.38.

<table>
<thead>
<tr>
<th>Tube</th>
<th>GE VCT®</th>
<th>Philips iMRC®</th>
<th>Varian MCS®</th>
<th>Siemens Straton®</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ø</td>
<td>238 mm</td>
<td>200 mm</td>
<td>200 mm</td>
<td>120 mm</td>
</tr>
<tr>
<td>Max. heat conduction</td>
<td>–</td>
<td>10 kW</td>
<td>–</td>
<td>50 kW</td>
</tr>
<tr>
<td>Max. heat radiation</td>
<td>15 kW</td>
<td>10 kW</td>
<td>17 kW</td>
<td>–</td>
</tr>
<tr>
<td>Anode heat storage capacity</td>
<td>6 MW</td>
<td>1.25 MW</td>
<td>5 MW</td>
<td>0.6 MW</td>
</tr>
<tr>
<td>CT power</td>
<td>140 kW</td>
<td>140 kW</td>
<td>100 kW</td>
<td>80 kW</td>
</tr>
</tbody>
</table>

FIGURE 2.34 Selection of targets for rotating anode tubes from different X-ray tube vendors for computed tomography.
The absolute focal spot temperature is not the only limiting parameter. Of course, other than for liquid metal anodes, the melting point of the target material of rotating anodes should never be exceeded. Thermomechanical stress in the focal spot rises steeply with the temperature gradient $\Delta T_{FS}$. Therefore, this gradient is often used as a second limiting factor during specification of the anode.

Given the benefit of a high melting temperature of the target material $T_{\text{max}}$, its atomic number $Z$, which is proportional to the conversion efficiency, heat conduction $\lambda$, mass density $\rho$, and specific heat capacity per mass unit $c_p$, target material for rotating anodes can be ranked by their figure of merit $Q = Z T_{\text{max}} (\lambda \rho c_p)^{1/2}$, see Oosterkamp (1948a). Under this premise, tungsten is the optimal converter material for a continuum spectrum of bremsstrahlung. This is also valid for stationary anodes and long exposure times, where the figure of merit becomes $Q_{\text{stationary-anode}} = Z T_{\text{max}} \lambda$.

For mammmography, however, the contribution of characteristic radiation is important as well as the interplay with the k-edge of the target material, which is used to narrow the spectrum. In mammography application with molybdenum or rhodium targets, maximal photon flux is traded for the benefits of the additional line spectrum and characteristic filtration.

The focal track erodes over time under the influence of the frequent temperature cycling. Despite a mild reduction of the melting temperature, a blend of tungsten with between 3 and 10% rhenium enhances ductility and extends the service time. A slight roughening pattern already marks the focal track after initial factory processing. A magnified close-up of the eroded target surface is shown in Figure 2.39. The initial micro-cracks partially relieve stress in the top layer. In the early stage of the operation of a tube, the X-ray conversion efficiency usually drops by 5 to 10%, stated for the standard 77 kV technique factors, measured with 25 mm additional aluminum filtration. As soon as an initial
Upon target erosion, the average depth of X-ray conversion increases from one to two micrometers to up to several dozen micrometers. Molten grains of tungsten may obstruct the free path of photons. Melting is enhanced once lateral cracks develop and the thermal contact to the depth of the bulk tungsten breaks down. All this causes beam hardening for tube voltages below the k-edge of tungsten, a lack of photons with energies above this k-edge for higher corresponding tube voltages, and an increase of the severity of the heel effect, as aforementioned.

### 2.4.4.3 Thermodynamics of the Bulk Anode

As Figure 2.36 demonstrates, the temperature in the anode is far from isotropic. During exposure, heat diffuses radially from a circumferential volume under the focal spot track to the cool center. Thermal diffusion from the outer circumference of the anode to the interior usually takes in the order of 10 seconds. Therefore, the energy input during a CT scan with a duration of several seconds will not be distributed entirely throughout the anode during this period. Heat stays at the outer circumference. The anode diameter has to be large enough to render a sufficiently large volume for heat storage. Rotating frame tubes are an exception. Due to their extreme heat conduction into the external cooling oil, their diameter can be reduced by about 40% without sacrificing too much performance. However, the thermal gradient in the depth direction of the anode is much higher than for anodes in a vacuum. Special care has to be taken to prevent these anodes from cracking, notably, as they are part of the vacuum housing. Thickness of the anode has second priority. Tube costs and mass rise with the anode diameter. It is therefore essential to specify the application as accurately as possible and to use optimal means for cooling, including heat conducting bearing systems, see below.

As a consequence of the thermal gradient, compressive circumferential stress develops under the focal spot track during exposure. During severe thermal stress, the material in and under the focal track is often hot enough to start creeping. It is squeezed outwardly, and therefore, partially releasing this stress while the thermal gradient to the interior is still prevailing. During cooling, a modified shape of the anode is frozen-in. Tensile stress remains after the phase of plastic deformation has ended. Upon cutting, a used non-segmented anode would cleave with an audible bang. The segmented anode in Figure 2.41 is essentially free of these issues. A small amount of fatigue may appear in the end-bores of the slits in the anode. This has to be considered by adapting the slit design to the thermal gradients in practical application.

### 2.4.5 Thermomechanics of the Bulk Anode

In order to set up the heat balance of the core component of a rotating anode, X-ray tube electron backscatter has to be considered, including multiple impacts on the anode as well as radiative...
and conductive cooling. As aforementioned, Figure 2.36 demonstrates that the temperature in the anode during exposure is all but isotropic. Heat diffuses from a circumferential torus-like volume to the cool center.

2.4.6 Cooling Channels

2.4.6.1 Backscattered Electrons

Avoiding heating is the most efficient way of “cooling.” As aforementioned in the context of the discussion on off-focal radiation, notably, high-Z materials like tungsten backscatter impinging electrons in high quantities. As these missing charge carriers are lost for X-ray production, they should at least not contribute to anode heating. An early attempt to carry the heat of scattered electrons away was made by Philips with the introduction of the Metalix® metal center section tube in 1929. A picture of the very similar rotating anode tube version Rotalix® is in Figure 7.16 and as a cut view in Figure 2.34. Historically, such a solution was already implemented by Bouwers (Philips) in the first rotating anode tube ever, the Philips Rotalix® tube, see Figure 12.18.

Other sub-components are prepared in the same way to efficiently radiate heat. Figure 2.2 shows the copper rotor of the glass tube covered with “black chrome.” Bouwers had already noticed the challenge in the late twenties: during a clinical patient sequence, heat radiation cooling alone of anodes suspended by ball bearings in a vacuum leaves a residual amount of heat in the anode. Heat radiation disappears with the fading glow of the anode. Figure 2.42 may demonstrate this finding in more detail.

During the first five minutes after exposure, and as long as the average anode temperature stays above 800°C, heat radiation is dominant. But then conduction through the liquid metal bearing into the surrounding cooling oil takes over. In this case, heat radiation nearly ceases after about 20 minutes. For typical clinical conditions, heat conduction is always dominant in the focal track of a rotating anode. This is what tubes with liquid

2.4.6.2 Heat Radiation

According to the Stefan–Boltzmann law, the temporal cooling rate, stated under the assumption of zero absolute ambient temperature and no reflection, amounts to

\[
\dot{T} = A \frac{\sigma \cdot \varepsilon}{\rho c_p} \frac{T^4}{V} \tag{2.3}
\]

\(\dot{T}\) denotes the time derivative of the temperature \(T\) of the body with a non-concave surface area \(A\), volume \(V\), mass density \(\rho\), and mass-related specific heat capacity \(c_p\cdot\sigma = 2\pi5k_B/(15\hbar c^2) = 5.67 \times 10^{-8} \text{ J m}^{-2} \text{ s}^{-1} \text{ K}^{-4}\) stands for the Stefan–Boltzmann constant of heat radiation, \(k_B\) the thermodynamic Boltzmann constant, \(\hbar\) Planck’s constant, and \(c\) the velocity of light. \(\varepsilon\) denotes the thermal emissivity of the body, which is the difference between albedo and unity. More details can be found in Behling (2016).

With the MCS® CT tubes, Varian optimized heat radiation from the rotating anode and maximized the emissivity by brazing a large grooved carbon back to the metal cup. Stationary cooling fins in thermal contact with the tube frame reach deeply into the grooves, see Figure 2.34. Historically, such a solution was already implemented by Bouwers (Philips) in the first rotating anode tube ever, the Philips Rotalix® tube, see Figure 12.18.

Table 2.5 provides a comprehensive overview of backscattering characteristics from the thermal perspective.
bearings are capable of providing. Not all vendors exploit this capacity to a full extent, however. Siemens added the final step and it nearly works without any heat radiation with their rotating frame tube Straton®. Cooling is based on heat conduction and collection of scattered electrons only. Figure 2.43 provides an overview of the various cooling concepts.

### 2.4.7 Temperatures in CT Application

The performance of tubes for CT is typically restricted by the thermal characteristics of the anode only, whilst for tubes for general radiography and interventional angiography also the capability of the cathode is limiting. Summarized, it is beneficial to maximize the anode diameter and cool by heat conduction. Such tubes deliver what they promise. A more detailed treatment of anode temperatures is provided in Behling (2016). It is a paradox that low-end CTs, with their limited coverage, require more capable tubes than high-end systems. Top-tier CTs, with enhanced gantry speed and larger coverage, consume less energy per patient but require higher focal spot loading to produce the required photons in less time.

#### 2.4.8 Temperatures in Interventional and General Radiography Applications

In interventional and general radiography applications, the focal spot temperature rise \( \Delta T_{FS} \) is the primary limiting factor. Energy input per patient is far smaller than for CT. Even CT-like procedures run in an angiography suite are less demanding. The large detectors make much better use of the generated photons than in
CT. Therefore, anodes for angiography tubes may be slim. The focal track speed is a key parameter for this application. When discussing large anodes, it is essential, however, to keep an eye on the momentum of inertia and the time to start the anode. Therefore, high-performance angiography tubes should always run continuously. The ideal solution is using a liquid metal bearing. High heat storage capacity alone is an inadequate quality parameter.

2.4.9 Thermal Performance Metric

2.4.9.1 Anode Heat Storage (Mega Heat Units) Abandoned

For decades, ball bearing technology and pure heat radiation cooling of anodes in glass tubes dominated the tube market. The quest to improve the workflow of CT has urged vendors to raise the heat storage capacity of rotating anodes time and again. Anode heat storage capacity and its unofficial unit, the Mega Heat Unit (MHU), became a synonym for performance. But, in view of newer technology, why store enthalpy which would not even embark on the anode but be taken away by backscattered electrons? Why store thermal energy when it can be better dissipated efficiently by heat conduction? The advent of liquid metal bearing technology toppled the outdated logic, as Figure 2.44 reveals. Rotating frame tubes perform well in CT, without the need to advertize any “Mega Heat Units.” Consequently, the IEC introduced a new standard 60613 in 2009 and put anode heat content (AHC) and all anode related Heat Units to rest.

The new edition three of the IEC 60613 from 2009 skipped discussing virtual heating charts and cooling curves, which the user is unable to validate. A detailed reasoning can be found in appendix A of the standard, see IEC 60613 (2009). For convenience, part of the text may be cited:

NOMINAL ANODE INPUT POWER: highest constant ANODE INPUT POWER that can be applied for a single X-RAY TUBE LOAD in a SPECIFIC LOADING TIME and under SPECIFIED conditions.

The nominal anode input power is presented as a function of exposure time in a chart like Figure 2.45. The following definition is useful for interventional and general radiography:

NOMINAL RADIOGRAPHIC ANODE INPUT POWER: NOMINAL ANODE INPUT POWER which can be applied for a single X-RAY TUBE LOAD with a LOADING TIME of 0.1 s and a CYCLE TIME of 1.0 min, for an indefinite number of cycles.

This is a fair approximation of clinical practice. The nominal radiographic anode input power is stated as a single value that characterizes the short time power that a tube can sustain for the production of a single exposure of 100 ms length every minute. This characterizes pulse performance as well as cooling capability and is therefore a very practical metric.

CT is special. Exposure times are typically several seconds long before the equipment is rearranged in about 10 minutes and the tube is allowed to cool down. According to this typical way of working, the IEC defined another practical metric, which allows for a reasonable comparison:

NOMINAL CT ANODE INPUT POWER: NOMINAL ANODE INPUT POWER which can be applied for a single X-RAY TUBE LOAD with a LOADING TIME of 4 s and a CYCLE TIME of 10 min, for an indefinite number of cycles.

1st patient okay. But, the glass tube will be damaged while exposing the 2nd patient (focal track erosion, risk of anode crack)

The metal ceramics tube with liquid bearing will well perform the entire exposure sequence

FIGURE 2.44 The metric anode heat storage capacity (AHC), stated in “Mega Heat Units” (MHU), is inadequate in view of modern tube technology. Compared are (a) an exemplary glass tube with graphite-backed anode and ball bearings, for which the metric was once defined, and (b) a metal center section tube with liquid bearing. Both are specified with 8 MHU. Tube (a) will be damaged by anode track erosion and possibly delamination of the metal–graphite compound after the second patient in the sequence has been examined. Due to better cooling and electron capturing, tube (b) will survive, although it has no better rating. Instead, the IEC 60613 ed. 4 has delivered a realistic metric, see text.
The above simple scalar value newly characterizes the practical capability of the CT tube instead of the hidden enthalpy stored in a tightly encapsulated invisible anode. The unit Mega Heat Units or cooling curves, which only the vendors are able to validate, have become obsolete. The new standard covers exposure performance as well as cooling performance. The loading is assumed to be repeated in a practical sequence, emulating a realistic average patient frequency. It reflects modern CT systems with typical exposure times. A more sophisticated alternative to the simple power figure is defined in the standards as well:

| CT Scan Power Index (CTSPI): characteristic of an X-ray Tube Assembly intended for use in Computed Tomography for a Specified range of Loading Times for single loadings, for a given Cycle Time, as follows: CTSPI = 1/t_{max} - t_{min} \int_{t_{min}}^{t_{max}} P(t) dt, where t_{max} is the upper limit of the Loading Time in seconds, t_{min} is the lower limit of the Loading Time in seconds, and P(t) is the function representing the Single Load Rating in kilowatts. |

Other parameters describe long-term cooling of the anode and tube housing assembly and the performance of the heat exchanger:

- X-ray Tube Assembly Input Power: mean power applied to an X-ray Tube Assembly for all purposes before, during, and after loading, including power applied to the stator of a rotating anode X-ray Tube, to the filament, and to any other device included in the X-ray Tube Assembly.
- Nominal Continuous Input Power: Specified highest X-ray Tube Assembly Input Power which can be applied to an X-ray Tube Assembly continuously.
- Continuous Anode Input Power: Specified highest Anode Input Power which can be applied to the Anode continuously.

### 2.4.9.2 Legacy Heating and Cooling Curves

As many product data sheets still phrase the matter in old terminology, this outdated language shall be briefly discussed.
Figure 2.45 is a historic set of heating and cooling curves defined according to the outdated edition two of the IEC standard 60613. It provides a simple numerical single-level heat integrating model according to

\[
\text{AHC}(t) = \text{AHC}_{\text{max}} \left[ \frac{T(t_0) - T_{\text{ambient}}}{T_{\text{max}} - T_{\text{ambient}}} \right] + \int_{t_0}^{t} [P(t) - P_{\text{cool}}(t)] \, dt
\]

where \( \text{AHC}_{\text{max}} \) represents the maximum anode heat content at highest anode temperature \( T_{\text{max}} \), assumed to be isotropic, \( T(t_0) \) its temperature at the starting time \( t_0 \), \( P \) the anode input power, \( P_{\text{cool}} \) the heat dissipated by the anode. For simplicity, invariance of the specific heat from temperature is assumed. “Anode” comprises all parts on anode potential and usually includes the rotor system. Implicitly, the tube frame is excluded, although it may be on “anode potential” in anode grounded tubes. Some readers may miss “Heat Units” (HU) in this context. Indeed, IEC relies on the SI system of units. It avoids the ambiguous unit HUs on purpose. As aforementioned, the accepted conversion factor between anode heat content in Joules and in Heat Units is valid only for legacy two-pulse high-voltage generators. However, it has also been common to use it for DC generators: 1 [Heat Unit] = \( \sqrt{2} \) [Joule].

Historically, heating and cooling charts served as a simple means to avoid overheating of tubes in sophisticated exposure schemes like cine and series exposures mixed with fluoroscopy runs. It may be instructive to interpret the use of such a chart, depicted in Figure 2.45. The anode of the tube is assumed to gain enthalpy (AHC) according to one of the heating curves. The AHC value may rise for the exposure time \( t_{\text{exp}} \) along the power curve which approximates the average energy supplied during \( t_{\text{exp}} \). This may comprise a single exposure or a series of X-ray pulses, for example, a cine run. The (single) cooling curve describes heat dissipation in terms of loss of enthalpy (heat content) over time and reaches from the maximum stated for the highest possible anode temperature to zero AHC at ambient temperature. The maximum should never be used for clinical use. IEC requested that the tube should still function after validation. But components may slightly deteriorate. The following shall explain the way users may simulate a sequence of exposures or runs of pulse series. For illustration, the graph assumes two heating cycles and a break in-between. The first heating starts with a cool tube at zero AHC and, with an exemplary 12 kW average load, may bring the AHC to an upper value of 1750 kJ (corresponding to about 2.5 Mega Heat Units) in four minutes. A cooling period, \( t_{\text{cool}} \), of seven minutes may follow, during which the AHC shrinks to about 600 kJ. The second phase of heating, with an average of 5 kW for three minutes, concludes the use of the tube with a final AHC of 1050 kJ, obviously without overheating it. None of the power curves the user followed during this exercise broke off. Nor was the maximum of the permissible maximal heat content ever exceeded. Nevertheless, key parameters of the procedure are out of control by the user. Unlike input power, which can be derived from generator readings, and exposure time, neither temperatures nor the AHC can be validated by the user. AHCs and cooling curves were stated by the manufacturers only.

2.4.9.3 Single Load Rating Chart

The next relevant data are part of actual accompanying documents for tubes. Figure 2.46 shows a current and power chart. The maximum permitted tube currents \( I_t \) are shown for loading times \( t_{\text{exp}} \) on the abscissa. The curves are labeled by the corresponding
tube voltage. Endpoints of the curves for tube voltages below the isowatt point are defined by the limited emission capability of the cathode. Therefore, a nomogram specifies the capability of an anode to be energized for a defined exposure time. But, at the same time, it describes the cathode performance of a rotating anode tube. Usually, this chart is written for a permitted long-term tube loading of 250 W, that is, a sequence of exposures or a continuous loading of this power. In order to push for larger power figures, some vendors state the power versus time chart for tubes which are assumed to be run cooler, for example, with a long-term average power input of only 20 W. This often does not reflect the clinical routine. The tube would not hold what the chart promises. Multiple sub-components contribute to the limits stated in the chart. The left (short-term exposure) values represent the capacity of the focal track and are limited by the focal spot temperature or the temperature gradient in the focal spot. Values for tens of seconds and beyond of exposure time are indicative of the anode size and the overall cooling performance.

The peculiarities of interventional X-ray imaging were pointed out before. In this application, pulse sequences for the production of cine movies or lengthy series of single shots for generating angiographic run-off patterns are blended into fluoroscopic pulse series, for example, for navigating a catheter. Modern X-ray systems are fully self-controlled. No radiographer is obliged to read series and cine data anymore before exposing a patient. Nevertheless, these data tables serve for benchmarking purposes and are part of the tube documentation. Details can be found in Behling (2016).

Although CT-like 3D-imaging is also a well-established part of interventional radiography, no dedicated way of specifying their capacity for this purpose has been devised yet for angiography tubes. Unfortunately, up to the time of writing, the new IEC term Nominal CT Anode Input Power is hardly stated for any of the available interventional angiography tubes. For pulsed operation, cine data may do. In addition, users may read performance parameters from single load rating charts, see Figure 2.46, to assess the capacity for continuous load operation of several seconds (CT scans).

### 2.5 Rotor Systems, Drives, and Vacuum Bearings

The use of bearings in a vacuum is a technical challenge. Hydrocarbon-based lubricants are not ultra-high vacuum compatible. Without further measures, absence of intermediate layers soon causes pitting and severe erosion of balls and raceways of ball bearings. Two classes of bearings have evolved over time: ball bearing systems with coated members and the advanced class of liquid metal lubricated journal bearings, also known as spiral groove bearings (SGB) or simply liquid bearings.

#### 2.5.1 Ball Bearing Systems

The vast majority of low- and mid-tier rotating anode X-ray tubes is equipped with ball bearings, as shown in Figure 2.47 for an exemplary tube for general radiography. Figure 2.48 depicts an assembly station for a large bearing for a CT tube. In both cases, special hardened steel balls are thin coated with lead, silver, or other substances, which separate steel from steel in the vacuum. Separating layers are typically a few hundred nanometers thick. Too thick a layer may create material clusters during operation and bearing noise. Noise and vibration have been fought over decades with quite some success. A historic solution from Philips for noise suppression was the radial spring suspension, as depicted in Figure 2.47. Improvements of compact ball bearing systems with matched members allowed the elaborate construction to be replaced by simpler concepts. Hertzian stress in the contact zones of bearing members remains an increasing challenge for the use of ball bearing tubes in high-speed CT systems. Bearing wear and tear is an ongoing issue, as Figure 2.49 reveals. Increasing size and inertia of members in large bearings causes higher relative roll speeds of the surfaces, which in turn enhances wear. Thus, more and more vendors turn to the scalable liquid bearing technology, see Figure 2.50 and the next paragraph.

#### 2.5.2 Spiral Groove (Liquid) Bearings

Following Philips, who introduced molybdenum-based heat conducting bearings in 1989, in the mid-nineties, Toshiba began...
commercializing steel-based spiral groove bearings, although without implementing significant heat conducting capability. Siemens introduced high-speed spiral groove bearings in the same period with compact anodes of 120 mm diameter in the Megalix® tube for angiography. In the meantime, all these vendors have also been offering the technology for CT application. GE followed recently as well. Another vendor is expected to work on this technology too.

The spiral groove bearing of a rotating anode X-ray tube comprises a system of four individual load carrying components. Two cylindrical radial bearings take radial forces and gyroscopic and other momenta, two flat bearings handle thrust loads. Figure 2.50 shows an example. A film of liquid metal, a eutectic system of gallium, indium, and tin, fills the gap between stationary and rotating members of about 20 µm. Upon rotation, members are separated by the liquid and generate a hydrodynamic overpressure in the liquid. In addition to capillary forces, fishbone-like structures of grooves help to keep the metal inside the bearing structure. Other means encapsulate the liquid and prevent leakage into the vacuum. Good heat and electric conduction of the structure are major benefits. Figure 2.51 represents the latest development, a liquid bearing which can be supported on both sides in order to guarantee enough stiffness and to withstand extreme centrifugal forces in CT. This bearing has been first used in the Philips iMRC® tube and later also in a similar design in the Siemens Vectron®. Extensive treatment of the theory of hydrodynamic bearings, historic challenges, and technical limitations can be found in Behling (2016).

**FIGURE 2.49** Worn-out high-performance bearing of a CT tube. The left raceway is still intact and the respective balls appear as new. The right subsystem shows strong erosion of all members.

**FIGURE 2.50** Members of a hydrodynamic liquid metal spiral groove bearing.
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2.5.3 Rotor Drive

Typically, a rotating radial magnetic field is coupled through a copper cylinder to generate driving torque. A close-up of a cut view of such a system is depicted in Figure 2.47. Eddy-currents arise in the copper cylinder, which in turn add an additional phase shifted magnetic field. A Lorentz-force in the rotor results, which acts tangentially and either drives the rotor when the external field rotates or brakes the spinning rotor if the external field is stationary, see Behling (2016).

2.5.4 Rotor Dynamics, Moment of Inertia, and Start-Up Time

Rotors resist changes of their state of rotation. Starting a polar revolution and raising the angular velocity $\omega$ requires torque $T_p$. Inertial mass $m$ relates, in this respect, to the polar moment of inertia $I_p$ given by integrating over the entire volume $V$ of the body

$$I_p = \int \rho(r) r^2 \, dV$$

where $r$ is the distance of the volume element $dV$ of integration from the center axis and $\rho(r)$ the local mass density. The angular velocity rises during acceleration with a constant central torque $T_c$ from $\omega(0) = 0$, as expected, with

$$\omega(t) = \frac{T_c}{I_p} t$$

One aspect to remember is that the moment of inertia steeply increases with diameter of the anode. A closer look to a typical glass tube with ball bearings in Figure 2.47 shows an anode disk of about constant thickness. Its moment of inertia grows with the forth power of its diameter. Ignoring the small contribution of the copper cylinder and assuming a homogeneous body of mass density $\rho$, constant thickness $h_{\text{anode}}$, and anode diameter $d_{\text{anode}}$, the momentum of inertia turns out to approximately obey a fourth power law with the diameter

$$I_p \approx \frac{\pi}{2} \rho \cdot h_{\text{anode}} \cdot d_{\text{anode}}^4$$

This has tremendous impact on the clinical workflow with ball bearing tubes, as the preparation time, and also the energy which has to be supplied to the tube housing assembly, rises. Assuming a given motor drive with driving torque $T_c$, the time $t_{\text{prep}}$ to start a ball bearing tube has to be increased by about 40% when the anode diameter grows by only 10%, as

$$t_{\text{prep}} = \omega I_p / T_c \approx \text{const} \cdot d_{\text{anode}}^4$$

In practice, start-up times range from about one second for a low-speed anode spinning at about 50 Hz to up to about five seconds for a heavy duty CT tube. Whilst for CT, where the entire gantry has to be accelerated before scanning, the start-up of the system is long enough to hide tube preparation, this does not hold for general radiography and interventional X-ray, where preparation times are of high importance. As long as the heat balance allows, a ball bearing tube with a small anode is therefore beneficial. Constantly rotating spiral groove bearings (SGB) solve the problem entirely. This is the reason why Philips could nearly double the anode diameter by introducing the SGB and even reduce $t_{\text{prep}}$. Another disadvantage of large anodes in air cooled ball bearing tubes arises from the fact that a larger share of the total available cooling power has to be reserved for starting and stopping the anode, which negatively impacts on the workflow.

2.5.5 Vibration and Noise

Rotors of X-ray tubes must be balanced after assembly and, in some cases, rebalanced after heating for the first time. Figure 2.52 shows exemplary balancing drills at the outer circumference of a graphite-backed anode of a tube for CT.

It is important not to run rotors of X-ray tubes with frequencies close to any of the intrinsic resonances of the tube, most importantly not at the main rotor resonance frequency. Unfortunately, the bearing suspension in X-ray tube housing assemblies can hardly be constructed stiff enough to avoid intrinsic resonances in the range of relevant rotor frequencies of high-speed tubes. The Philips iMR® tube in Figure 2.25, which has a dual suspended spiral groove bearing in a stiff metal–ceramics frame is an exception. Intrinsic resonances of most other tubes are found to be between 3000 and 9000 rpm.

High-speed tubes usually operate at frequencies above the main intrinsic resonance, that is, hyper-critical. The tubes shown in Figures 2.30 and 2.47 were equipped with radial spring support of the rotor in order to reduce the intrinsic resonance frequency. Another way to manage this issue is reduction of the stiffness of the tube support in the tube housing. The downside of these measures is an increasing dependency of the focal spot position on the amount and direction of external forces like gravity and centrifugal forces. A stiff design, as depicted in Figure 2.25, is therefore beneficial in view of the accuracy of the definition of the focal spot in the system.

It is not advised to let rotors with ball bearings coast from speeds above resonance without applying an enforced magnetic
brake. Such rotors, in particular those without internal axial spring pre-load, may feature coast times of an hour or more. The rotor frequency may then dwell for a long time in or near the intrinsic resonance with a negative impact on the bearing life.

2.5.6 Gyroscopic Momenta

When tilting the spinning rotor of an X-ray tube with a polar momentum of inertia $I_p$ and rotor speed $\omega$ with a speed of enforced precession $\omega_g$, a resulting gyroscopic momentum $M_g$, will amount to

$$M_g = I_p \omega \cdot \omega_g$$

The direction of the angular momentum of the tube rotor in CT systems will stay unchanged under rotation of the gantry and no gyroscopic momenta arise, not even when the gantry is in a tilted position. Gyroscopic effects gain importance in interventional X-ray imaging where the C-arm angulates the tube in a co-planar manner. Current angiography tubes typically sustain precession speeds of roll motion of about 30 degrees per second.

2.6 Manufacturing of X-ray Tubes

2.6.1 Cleanliness

Cleanliness of all sub-components is key for high-voltage stability, long bearing life, and stable electron emission. This holds, in particular, for the interior components of the tube insert. Components for X-ray tubes are generally all but commodities and often require use of exotic materials and manufacturing processes in house and at the supplier. Before final assembly, sub-components are subject to multiple mechanical, chemical, galvanic, and thermal treatments. Compatibility with ultra-high vacuum requires extensive surface cleaning and degassing. A diffusion pumped furnace for this purpose is shown in Figure 2.53. In addition to parts heading to the vacuum side of the tube insert, the insulating oil must also be free of particles and dry enough to prevent accumulation of conducting filaments of micro-particles under the influence of the electric field. Quality manufacturers have invested heavily in dust-free assembly rooms with controlled humidity and airflow, as shown in Figure 2.54. Tube components are assembled under laminar airflow. Parts locks decouple polluted and clean areas. Sub-components usually undergo extra ultrasonic cleaning steps before being inferred from buffer stock into a controlled assembly area.

2.6.2 Production Lines

Figures 2.55 through 2.57 illustrate steps in lines for the production of metal ceramics tubes and for glass tubes. After thermal degassing of individual sub-components and their assembly, tubes are typically first evacuated to a basic residual gas pressure of about $10^{-7}$ Pa by baking the entire vacuum tube insert. Individual subsequent heating of cathode and anode beyond the limits of operation in clinical practice ensure proper vacuum levels. After pinch-off, see Figure 2.58, the tube inserts are placed either into an oil bath for high-voltage conditioning or directly into the lead shield to prepare for the same. The entire tube housing assembly is then filled with degassed and dry oil. Yet when a tube assembly is operated, vacuum discharges will occur for tube voltages above half of the nominal value. Therefore, a high-voltage conditioning process to about 20% overvoltage follows in an attempt to destroy troublesome electrode microstructures and to further reduce the residual gas background. As there is a risk of fatal errors during high-voltage conditioning, some manufacturers prefer to condition individual tube inserts in oil vessels before assembly into the tube housing. Electric stress and temperatures are driven above the limits for clinical use to ensure reliable operation in practice.

2.6.3 Break-In, High-Voltage Conditioning

Particulates, oxide layers, and other residual pollution on electrodes may cause reduction of the local work function or enhanced electrical field strength, for example, at the apex of protrusions. The interface between insulating particles and metallic electrodes may form triple points, which originate from differences in the electric susceptibility, polarization, and
charging from impacting charge carriers or photo-ionization. Upon the action of elevated electric field strength, these weak spots may constitute origins of field emitted electrons. Electron current may result in local heating, vapor, ion, and plasma generation and eventually short circuiting of the vacuum gap. Stability of X-ray tubes under the application of high electric fields between adjacent electrodes cannot usually be achieved by cleaning and other surface treatment only. In addition to thermal treatment, conditioning processes are applied, which stimulate the electrical explosion of any local instabilities on purpose. A high voltage is supplied, causing increasing field strength, supported by thermal excitation, gas production, and sophisticated cycling of the technique factors applied. Between several dozen and up to about a thousand discharge events may occur under controlled damped supply of power until all relevant critical surface features are safely destroyed and the tube
is able to operate at the specified conditions. In some cases, a stable situation is not achievable, where for example, the vacuum level is insufficient or if parts of the tube, like insulators or bearings, released debris. Such a tube would be scrapped or may be recycled and undergo a repair and second processing cycle. Figure 2.59 shows a glass tube during exhaust and electron bombardment of the anode. The figure nicely depicts a fluorescent glass wall next to the anode rotor, excited by impact of electrons which are backscattered from the anode. Electric current flows from the cathode to the focal spot on the anode disk, then in the form of backscattered electrons onto the glass wall and, supported by ion impact, again as backscattered electrons from there back to the only positive electrode, the anode. Concurrence of excessive gas pressure and inner surface coating of the glass wall may cause narrowing of the current paths, local overheating, and tube implosion. A “frosted” glass tube is shown in Figure 2.60. The inner glass wall was roughened by chemical etching for better resistance against creeping charges. Figure 2.61 depicts a tube insert during high-voltage conditioning in an oil container.
2.6.4 Final Testing

Figure 2.62 shows an exemplary test position for leakage radiation inspection. Leakage radiation is among those parameters which are extremely tightly monitored during production of each tube housing assembly to avoid any safety hazard for the customer. Other tests, like implosion or over-temperature tests are mandatory only during the development phase of a new type of X-ray source. Safety margins ensure that manufacturing tolerances are properly considered. Figure 2.63 gives an impression of a manufacturing facility for high-performance tubes. X-ray tube housing assemblies are waiting in line for final high-voltage testing, focal spot measurement, and other measurements.

2.6.5 Process-Oriented versus Assembly-Oriented Production

Although experience is growing with every day of production, the very complex physics inside an X-ray tube would not allow tracking of the vast entirety of physical parameters which are relevant for performance. This may surprise the reader. However, the influence of inevitable changes of the raw material and rare occurrences of adverse conditions render it impossible to predict all the measurable events. For example, vacuum discharges still appear in a “stochastic” manner. Although rare, they may impact clinical practice. Their exact timing and frequency is unpredictable. It is impossible to totally scrutinize all electrode surfaces down to the atomic level. A sound guideline is maintaining
material parameters as stably as feasible, and always repeating the same set of processes, gained from experience. Drawings and specifications of sub-parts are detailed. It is not sufficient to simply categorize the known key parameters of sub-components, specify them numerically, and check for compliance. Many aspects are hidden and may surface only after long periods of production, if at all. After initial specification and life testing of a novel tube type, quality is guaranteed by repetition and correcting for deviations. Steady process improvement is a must. Thus, to safely ensure the performance, processes of tube production have to be repeated with the highest possible accuracy. Tube production is thus called process-oriented. This is different from assembly of electronic equipment like high-voltage generators. The assembly process is, in their case, characterized by clear definition of a huge number of interfaces and sub-components and ideally predictable behavior. The high-voltage tank of an X-ray source, with its complex electrical interactions, is an exception from this rule, see Chapter 3 on high-voltage generators.

2.6.6 Production Yield

Among other factors, the success of manufacturing high-quality X-ray tubes is dependent on staff experience, staff retention, operations management, sophistication of the manufacturing processes and their steady improvement, number of product platforms to be maintained, quality of tools and other equipment. Decades of learning have enabled the production yield in world-class lines to exceed 97%, even for complicated top-performance tubes. In the eighties, typical scrap rates were an order of magnitude higher and occasionally peaked at 100%. Despite a high degree of automation, the quality of manual labor, as illustrated in Figures 2.54 and 2.55, is still a key success factor for every vendor. A small number of tubes have to undergo multiple loops during processing and remediation cycles, for example, when they fail the final high-voltage test. High factory yield is often indicative of good performance in the clinic as well.

2.7 Installation and Service

2.7.1 Reconditioning

Over time, insulators discharge and depolarize. Gas molecules slowly, but steadily, diffuse from hidden gas traps into the free vacuum space and may adhere to former clean electrode surfaces. The focal spot track, which had been cleaned by thermal desorption of gas molecules, is again becoming polluted over time. Loose particles may be released. All these factors contribute to the destabilization of the once-achieved stable surface condition of the electrodes of a tube. New centers of field emission emerge. During first application of high voltage after an extensive period of storage and transport, electrical breakdown may occur at low voltage levels. It is therefore advised to begin operating a newly installed X-ray source carefully and start raising the tube voltage from levels of only about half the nominal. The characteristic time for the above described processes to happen is not well-defined. A newly installed X-ray tube should be carefully reconditioned according to the break-in procedure suggested by the manufacturer. Tubes should be returned to the manufacturer for more extensive reconditioning after, for example, about half a year of shelf time.

2.7.2 Warm-Up

A short warm-up and reconditioning procedure should generally be applied after a pause of operation of several hours to bring the sub-components of the tube close to the thermal conditions during initial calibration of the system. Electron emission depends on the cathode temperature. Matching both components, that is, the adaptation of the generator and tube, should be performed in pre-heated conditions.
2.8 Tube Replacement and Recycling

2.8.1 Average Tube Life Time and Warranty

Figure 2.64 illustrates a selection of typical tube failures. As long as dominant epidemic failures are absent, a broad spectrum of failure modes usually contributes to the loss of tubes. Figure 2.65 presents a close-up of punctured ceramics after high-voltage discharge. Reconditioning is not possible in such a case, other than for most of the metal-to-metal discharges, the remnants of which are shown in Figure 2.66. Secondary effects may kill a tube as well. Figure 2.67 shows a burned high-voltage plug. Even if this event occurs at the generator terminal and the tube seems operational, the asymmetry of the high voltage may have caused damage of the tube.

Objective public data on average tube service life are scarce. Anecdotic and internal data suggest significant differences between vendors. Erdi (2013) reports on 50 tube replacements in 10 GE LightSpeed® and three GE VCT® scanners. This paper reveals a broad temporal and operational distribution of tube life. GE Performix Ultra® tubes for one of the two system types

FIGURE 2.64  Typical wear and tear patterns of returned medical X-ray tubes. (a) Glass erosion from high electric currents across the glass frame after coating by tungsten vapor. (b) Anode track erosion after long service life. (c) Cracked rotating anode. (d) Evaporated and eventually molten hot spot of an electron emitter coil. (e) Heat exchanger oil pump failure caused overheating and collapse of the tube frame. (f) Foot-point craters from vacuum-discharges.

FIGURE 2.65  Punctured cathode ceramics.

FIGURE 2.66  Remnants of severe tube arcing in a bi-polar metal center section tube (anode removed).
under investigation lasted on average $1.6 \pm 1$ years, Performix Pro® tubes in the other series of systems $1.9 \pm 0.8$ years. The total electric charge conducted during tube life differed by more than an order of magnitude between good and bad individuals of the Performix Ultra® series, namely $16.7–239.9$ kA with a mean value of $81.0 \pm 45.4$ kA. Performix Pro® tubes conducted $18.5$ up to $61.4$ kA with an average of $44.6 \pm 25.8$ kA. Some of the Performix Ultra® tubes did not survive the warranty period granted by the manufacturer. This is not necessarily an indicator of bad quality. Warranty terms often serve the purpose of insurance. Operating a medical X-ray tube is commercially risky. Buffering by a vendor guarantee or a service contract, which includes tube exchange, is advised. Whilst full warranty ends as contracted, pro-rated warranty reduces the potential refund from the vendor over time or cumulative amount of use in operational units, see Figure 2.68.

### 2.8.2 Recycling

Recycling of the tube housing has been common practice since the invention of this enclosure of the fragile tube. In addition to this kind of re-use, the introduction of metal–ceramics technology and, in particular, long-living liquid bearings allows qualified manufacturers to also recycle sub-components of tube inserts. After dismantling and exhaustive inspection, recycled sub-components can be re-used for new equipment to protect the environment and to save costs. Figure 2.69 depicts an example. Once a high-voltage insulator has survived hundreds of hours under the stress of high voltage and electron bombardment, the risk of puncture due to an intrinsic flaw is minimal. Philips has pioneered this green strategy, which customers and regulatory bodies have been blessing. Performance and tube life are guaranteed. Other than for rugged metal–ceramics tubes, the insulating capability of a glass frame is often irreversibly damaged after years of service, see Figure 2.64. Debris from ball bearings, see Figure 2.49, are polluting other sub-components. Thus, worn-out glass tubes with ball bearings usually have to be scrapped.

### 2.9 Value Engineering and Tube Costs

Figure 2.70 shows a cost breakdown structure derived for an average high-end rotating anode tube for CT. Anode size dominates the calculation. System developers should very carefully minimize their requirements in case a selected tube platform seems to come to its limits and they are tempted to suggest the next higher tier be used. Can the gantry speed be reduced for selected critical use cases? Can detector coverage be extended in a default application protocol to save energy input and anode costs? Is a larger focal spot acceptable? From a different perspective, Figure 2.71 demonstrates the impact of the planned clinical application on tube price. As aforementioned in the context

![Burned-out cable plug after false assembly.](image1)

![Careful dismantling for the recycling of an anode of a high-performance tube for computed tomography.](image2)
of rotor drives, larger anodes are not always better performing. A 90 mm diameter anode seems an excellent trade-off for general radiography, which balances minimal start-up time and rotor driving power with heat dissipation, in particular, as image de-noising has much improved over the years. Larger anodes would be more expensive and may even hamper the workflow. Figure 2.72 illustrates the typical costs of specific additional functionality.

REFERENCES


