3

X-ray Generators

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The historic term “generator” does not exactly reflect the function of this device. If we ignore niche products utilizing pyroelectric (see Section I, Chapter 6 of this book), as produced by the company Amptek, Bedford, MA, USA, piezoelectric, see Gall et al. (2013), ferroelectric, see Altgilbers (2009), or triboelectric effects, see Hird et al. (2011), to generate high voltage with low current, typical medical diagnostic high-voltage supplies do not per se generate any electrical energy. They rather convert AC mains power to high-voltage DC output, and usually augment the function of X-ray tubes with a variety of auxiliary signals and electrical supply. In this sense, X-ray generators are intelligent voltage step-up and current step-down electronics, delivering the required tube voltage for the generation of X-rays and other power.

A comprehensive treatment of the vacuum electronics and the power electronics part of X-ray generation can be found in Behling (2016) and in Chapter 7 of this work. High-voltage engineering is treated, for example, in Rizk (2014). Basic background on medical generators can also be found, for example, in Rossi et al. (1985), Krestel (1990), and in Behling (2016), Chapter 8.

3.1 Basic Functionality of the X-ray Generator

Radiologists often associate the “generator” with its user interface, as shown in Figure 3.1, which often comprises a major part of the user interface of a general purpose X-ray system (see Section II, Chapter 26 of this book). Figure 3.2 sketches the overall functionality of an exemplary diagnostic X-ray generator. Its main functions are delivering:

- Between 20 and 150 kV tube voltage and up to 2 × 120 kW of electrical DC power
  - With mono- or bi-polar high voltage, cathode charged and/or anode charged
  - With constant or slowly or fast varying tube voltage (“kVp-switching”)
  - With tube discharge damping and high-voltage recovery after tube “arching”
- Heating current on cathode potential for the thermionic electron emitters
- Grid voltage supply added to cathode potential for grid switched tubes
- Voltage supply added to cathode potential for electrostatic beam deflection and emission control
- Currents to feed magnetic quadrupoles and dipoles for tubes with magnetic focusing and/or focal spot deflection
- Stator supply for the motor of rotating anode tubes
- Tube voltage and tube current measurement
- and more
It:

- Is the interface between
  - User and system in radiographic equipment
  - Tube and X-ray system
- May control, depending on the use case
  - Total dose and dose rate
  - Tube current
  - Tube voltage
- Provides safety functions like
  - Thermal monitoring of the tube
  - Arcing switch-off
- Provides service functions including
  - (Remote) error logging and display
  - Application logging

### 3.2 High-Voltage Train

The schematic of Figure 3.3 sketches the basic electronic circuitry and illustrates major components of a modern medical diagnostic high-voltage generator. Electrical energy comes in with 50 or 60 Hz AC single- or three-phase mains of between 380 and 480 V and is converted to between 20 kV (mammography) and 150 kV nearly DC tube voltage. Different mains conditions on site are frequently managed by transformer-based power distribution units (PDUs). Coming from the left, an adapter (1) matches local mains conditions with the requirements of the power conversion unit of the high-voltage generator and the overall X-ray system. Further to the right, a combination of diode and smoothing capacitor rectifies the AC mains input, filters the voltage, and generates an intermediate DC rail voltage of typically between 550 and 750 volts. Metal-oxide-semiconductor field-effect transistors (MOSFETs) or insulated-gate bipolar transistors (IGBTs) (2) chop the rail voltage and feed resonance elements (3) with AC. The transformer (4) converts high currents driven by moderate AC voltages on the primary side to magnetic inductance in its ferromagnetic core, which its windings enclose, see for example Bushberg et al. (2012). The core penetrates a higher number of secondary windings, isolated from ground. High AC voltage is induced proportional to the temporal change of the inductance and the number of turns. Reduced inductance may be compensated by higher frequency. Enhancing the switching frequency...
is, therefore, improving compactness. Figure 3.4 illustrates the advancement by the transition from 18 kHz switching frequency to more than 100 kHz high-voltage supplies.

The exemplary interior of the larger unit to the right of Figure 3.4 is shown in Figure 3.5, a high-voltage tank of a bipolar generator for interventional diagnostics. The high-voltage transformer is surrounded by two towers, which integrate damping elements and resistive voltage dividers in the front of the left picture, which supply the control circuitry with the actual value of the high voltage. Filament heating transformers, which transfer AC current to the negative high-voltage potential, are in the center of the right picture.

High-voltage generators from different vendors can be differentiated by the individual design goals, the tradition and specific competence of the development team, the local specifics of the mains grid, and the application they are intended for. For example, some vendors prefer molded high-voltage plastics insulation in the high-voltage tank over oil. Others aim for minimal secondary capacitance for fast high-voltage switching (kVp-switching). Suppliers for the open market of original equipment manufacturers (OEMs) try to achieve an utmost degree of compatibility with systems from different vendors. This may affect the polarity offered (bi-polarity, mono-polarity) and—most importantly—the electrical interfaces.

It may be illustrative to show a few alternative architectures, which differ by the inverter frequency used. Figure 3.6 depicts a system comprising fast switching MOSFETs. These transistors allow for elevated switching speed, but impose limited rail voltage. Thus, a voltage step-down/current step-up unit (Buck Converter) is required which limits the DC input of the converter to less than 380 volts. An alternative is shown in Figure 3.7: unlike MOSFETs, this architecture works with IGBTs which allow feeding of the rectified and smoothened mains input directly into the inverter. The price to pay is lower inverter frequency, a larger high-voltage transformer, and larger output capacitance to reduce high-voltage ripple. Other alternatives can be found in the high-voltage architecture. Figure 3.3 depicts a sample four-stage rectifier cascade. Some vendors prefer a series of stacked secondary transformer–diode–capacitor sub-units instead.

In addition to tube voltage pulsing, tube current adjustment, steering tube rotor, electron emitter temperature, etc., modern generators also control the reaction on tube arcing. A discharge in the tube side is sensed as a negative voltage surge or an excessive current peak arriving at its output terminal. The control unit switches all power transistors off, waits for a cooling period of the order of several hundred microseconds, necessary to settle local gas and vapor bursts, and attempts to ramp-up the voltage again. Some implementations allow for many break-down
events to occur per second before an exposure is aborted due to lack of adequate X-ray intensity. A bad X-ray tube may exhibit reasonable performance when attached to an intelligent generator. Simpler systems use passive components such as resistors, diodes, and capacitor-inductance combinations, for example as described in the European patent EP592164B1.

Length and capacitance of the high-voltage cable is important in this context, as it stores electrical energy, $E_{\text{discharge}}$. In a
unipolar high-voltage design the energy released during a full discharge is

$$E_{\text{discharge}} = \frac{1}{2} V_t^2 \cdot C_{h/v}$$

where $V_t$ is the tube voltage and $C_{h/v}$ the total electrical capacitance in the high-voltage circuitry, which consists of the generator smoothing capacitance, cable capacitance and capacitance of the tube assembly. Resistive damping elements disconnect the internal generator smoothing capacitance from the usually lower cable capacitance. The energy which stresses tube electrodes during rather short vacuum discharges is, therefore, limited. In a bi-polar system, the “local” energy stored in the electrical capacitance close to the tube, that is, in the cable and in the tube itself, which is released in a single vacuum discharge, is only half of the above value, as the tube voltage splits into two halves and in each of the two branches one-quarter of the energy is stored. Cables in computed tomography (CT) systems are short, ca. 0.5–2 meters long. However, cable lengths of more than 40 meters have become common in many installations of interventional angiography and hybrid interventional and surgery systems. Discharges of electric charge from long cables may cause destruction of electrodes or insulators inside the tube, if not handled properly. Countermeasures may comprise resistive cable-based damping, special cooling periods during arc-ride through the generator, or dedicated processing of the tube during high-voltage breakdown, see Behling (2016). Enhancing the switching frequency with modern generators helps in reducing smoothing capacitance and the energy released during such kinds of events.

3.3 Magnetic Slip Rings

Computed tomography with common helical scanning introduces a technical challenge. There are several ways to transfer power from the stationary to the rotary side of the gantry in a CT system. The most common method employs mechanical power slip rings and brushes to transfer DC rail voltage, which is then converted to high-frequency AC power in a rotating power converter. Magnetic coupling is an appealing alternative, for example offered by Analogic, Peabody, MA, USA and introduced by General Electric (GE), Waukesha, WI, USA in systems for baggage inspection first and then diagnostic imaging. In one of the
solutions the intermediate rail voltage is chopped on the stationary side. AC is magnetically transmitted and directly drives the high-voltage transformer with high frequency. Vendors pursue different strategies to optimize the overall architecture in view of power rating, efficiency, costs, space consumption, versatility, risks of electromagnetic compatibility, and durability.

### 3.4 Dual Energy Imaging

Since 2008, GE has marketed spectral sensitive X-ray imaging in CT (for dual energy CT, see Section III, Chapter 39 of this book). It is based on ramping the tube voltage up and down, with steep rise times also referred to as fast “kVp-switching” (peak tube voltage switching). Projections are taken with low tube voltage, for example 80 kV, intermediate voltage during transients, and with high tube voltage, for example 140 kV. The spectrum of the bremsstrahlung from the X-ray tube is being altered in a sub-millisecond manner. This requires the high-voltage chain to pump charge into the smoothing and cable capacitance during ramp-up and withdraw this charge through the tube afterwards during ramp-down. Transition times as low as several hundred microseconds can be achieved, depending on the selected tube current, the current delivery of the power chain, and other means of discharge in the high-voltage generator. Recently, GE could shorten the transition times further with a new generation of generators for CT. Of particular interest are the total output capacitance of the high-voltage chain, the cable length, and the maximal current which the inverter can feed to the high-voltage transformer. As the tube current usually diminishes with lowering of the tube voltage when the cathode operates partly space charge limited, special measures are advised to actively steer the cathode of the tube with biased electrodes in parallel with altering the tube voltage, in order to fully employ the thermal capability of the focal spot.

### 3.5 Sensing Tube Voltage and Current

Feed-back loops are essential for controlling tube voltage and tube current. As outlined in Chapters 1 and 2, see also Behling (2016), the tube voltage determines the image contrast, whereas the tube current specifies the photon flux from the X-ray source.

The two towers of resistive high-voltage dividers, shown in the left picture of Figure 3.5, are capacitive balanced to achieve a bandwidth in the range of several hundred kilohertz and feed the control unit with a signal which represents the tube voltage. Figure 3.8 illustrates the basic electrical schematics. High bandwidth is necessary for fast control of the tube voltage in high-frequency generators and early detection of signs of vacuum discharges in the X-ray tube. Since the tube voltage has a direct impact on the dose delivered to the patient, the accuracy of the delivered tube voltage must be highly accurate.

The tube current, the second important technique factor, may be monitored with less precision and frequency bandwidth. Simplified exemplary current measurement circuitry is shown in Figure 3.8. Signals are fed back to the drivers, which control the switching scheme of the inverter. Sub-picture (8) of Figure 3.3 shows a sample control board, realized by field programmable gate arrays, digital signal processors, CPUs, etc.

The tube current equals the cathode current in most cases. Field emission from cathode to anode or cathode to ground usually amounts to much less than a hundred microamperes at highest tube voltage and drops significantly with tube voltage. The tube current sensing circuitry has to be adapted accordingly. Normally, that is, when thermionic emission with thermal emission control is used, electrons emerging from the emitter in the cathode hit the anode first. Thus, they contribute to X-ray production; 40%–60% of these are backscattered and may partly land on the tube frame and on areas of the anode outside the focal spot. Thus, the tube current returns completely through the earth terminal.

![FIGURE 3.8 Typical measurement circuits for tube voltage and tube current for a bi-polar high-voltage generator.](image-url)
of the source of negative high voltage, for example the negative
branch of a bi-polar generator. The current signal includes all
charging currents for the capacitance of the high-voltage cable,
tube assembly, and parasitic capacitance. The tube current sig-
nal has to be extracted by sophisticated electronic means from
substantial electrical “noise.” Transitions of tube voltage have to
be considered, for example during switch-on, fast dose control,
or high-voltage pulsing. The signal is then cleared by averaging
with low-pass filters to deliver a stable current reading. In addi-
tion, the measurement divider current has to be considered as
does the temporal dynamics, which requires a capacitive com-
ensation of the cable capacitance. The dynamic range has to
cover four orders of magnitude of current.

3.5.1 Multiple Tubes

For cost reasons, legacy radiographic generators were often con-
figured to serve two X-ray tubes. Figure 3.9 shows a high-voltage
switch for this purpose.

3.6 Energy Quantization

Using resonance converters, switching MOSFETs or IGBTs, as
shown in Figure 3.3, are initiated during zero-crossings of the
secondary voltage to avoid overheating of the semiconductor
die. It is not advised to quench high current at elevated voltage
levels and arbitrary points in time. Series-resonance converters
with half or full bridge switches have proved to be very energy
efficient. Energy efficiency translates to low operating tempera-
tures and long service life. For this architecture, the minimum
period of energy supply is half of the resonance period of the
power inverter circuit. Thus, the minimum energy transmitted
is the temporarily stored energy in the resonance elements. As a
negative consequence, any energy transfer to the tube is quantized.
Several degrees of freedom to control the tube voltage are avail-
able, such as pulse width, frequency, and so on. The inverter may
be in an off state during several resonance periods if the tube
is to be operated with low power. Consequently, the amount of
ripple changes when compared with full power operation. In case
of pulse-width modulation, the ripple can be reduced with power.
Multiple power ranges of operation may exist in an attempt to
minimize the undesired voltage ripple. The control scheme may
be adapted according to the desired ripple application.

3.7 Voltage Ripple

As it tends to worsen the patient dose penalty for a required con-
trast-to-noise relation in the X-ray image, voltage ripple is unde-
sired. Due to the energy quantization outlined in the previous
section and the use of sophisticated power control schemes, volt-
age ripple is often a complicated function of selected technique
factors, namely tube voltage and tube current. IEC 61676 defines
voltage ripple as the ratio between peak-to-peak voltage alteration
and maximum voltage. On purpose, the maximum (“kV-peak”)
is taken as a reference to define the X-ray spectrum, because the
resulting X-ray intensity at the detector downstream of the patient
after beam hardening is a strongly non-linearly rising function of
the tube voltage. Figure 3.10 may illustrate the strong impact of
tube voltage on the X-ray intensity, even at the entrance side of the
patient, which is relevant for the consideration of the skin dose.
The result of a simulation of the X-ray intensity for a hypotheti-
cal single-phase generator without rectifier—as used by Conrad
Roentgen during his first experiments—is shown as a reference.
Clearly, the diode characteristics of the bremsstrahlung source is
visible as is the dominance of the peak voltage for dose calcula-
tions. The maximum tube voltage reached during a voltage cycle
determines the total photon flux to a large extent. For the simulation
of Figure 3.10 the X-ray tube is assumed to operate a thermionic
space charge limited cathode. According to Child–Langmuir’s

![FIGURE 3.9](image_url) A legacy dual tube high-voltage switch cut open.

![FIGURE 3.10](image_url) Voltage ripple and X-ray dose during production of brems-
strahlung, with a hypothetical legacy single-phase system mains-driven gen-
erator without rectifying and smoothing means. The AC voltage phase is
shown on the abscissa. The ordinate indicates tube voltage and X-ray inten-
sity in arbitrary units.
law, the tube current $I_t$ is then proportional to $V_t^{3/2}$, where $V_t$ represents the tube voltage. The X-ray intensity from the source is then approximately proportional to $V_t^{1.5} \cdot V_t^{2} = V_t^{3.5}$. The X-ray intensity at the detector downstream of an exemplary patient of 300 mm water equivalent has an even stronger dependency on the tube voltage and is approximately proportional to $V_t^{5.5}$.

In modern generators, the absolute value of the ripple typically amounts to a single digit percentage of the tube voltage. For accurate dose and contrast calculations, the exact waveform for the technique factors applied has to be known. The amount of ripple in terms of a single figure would not suffice. More details can be found in Behling (2016).

### 3.8 Power Rating

Figure 3.11 depicts a power block of a modern high-frequency generator for CT, which delivers 60 kW electrical power to the X-ray tube, with a voltage ripple of less than 4%. The entire power supply of the compatible Philips iCT® system comprises two of these blocks and delivers 120 kW for 4 seconds every 10 minutes, which matches the compatible iMRC® tube. Tube and generator fit well. This is a most reasonable and modern specification method. Other metrics, such as nominal anode input power (for 0.1 second loading) or stating the “generator power” alone, would introduce ambiguity. Specification in practical terms following the new third edition of the IEC standard is the best way of comparing competing offerings.

The achievable power rating is not only dependent on the components used, for example for inverter, high-voltage transformer, and diodes, but also on the mains conditions on site. The 480 V three-phase AC mains allows for higher primary DC rail voltage (800 V DC) for the inverter than the 380 V AC mains (rectified to about 600 V DC). A generator optimized for 480 V AC has to conduct less current through the primary rectifying diodes and the inverter to deliver the required power at the high-voltage terminal. Vendors may use this degree of freedom to simplify and cost reduce the equipment. Generators, which usually rely on 480 V AC grids, would, therefore, need an expensive mains transformer interface to work optimally in other places.

### 3.9 Current Rating

On top of total power, the tube current which a generator can deliver under the given mains conditions and relevant tube voltages is an important characteristic. In addition to the considerations on power rating (see the previous paragraph), high tube current capacity requires well dimensioned high-voltage circuitry, see Figure 3.12. In particular, rectifier diodes, push-capacitors in the high-voltage cascade, smoothing capacitors, and the damping resistor must be able to sustain high tube currents and at the same time reduce the voltage ripple to an acceptable level, even when the tube discharges fast with high current.
3.10 Matching Generator and Tube

Commercial system specifications are sometimes misleading. Stating "generator power" alone for a CT system may raise the suspicion that other components might not match well with this device. The capacity of current delivery, speed of power control, voltage ripple under various mains conditions and technique factors demanded, electromagnetic compatibility, auxiliary functionality, safety, back-up and service functionality, hazard prevention, arc suppression, tasks adopted as part of the systems architecture, and more count, too. It is beneficial to have high-performance generators and tubes developed by a single R&D team. The complexity of the interface between both components has grown tremendously over time. Management of vacuum discharges between electrodes of the X-ray tube, magnetic focusing, electron beam deflection, and thermal management of the X-ray tube and the generator are only a few aspects to consider.

As mentioned before, power ratings and capabilities of current delivery of the cathode of the tube and the secondary components of the generator have to match well. Whilst the performance of the tube is primarily limited by the anode, namely the thermal balance of the focal track, cooling rate of the anode, and space charge effects in the cathode, the generator has slightly different characteristics. As outlined above, generator power and current ratings are defined by the losses in generator components and the current carrying capacity, for example of diodes. Saturation of the magnetic core of the transformer is another aspect to consider, as well as long-term heating and margins for transients of voltage and current. Simplified rating charts usually state maximal generator current and output power. Figure 3.13 illustrates the performance of two sample X-ray segments. Figure 3.13a represents an acceptable match. Figure 3.13b demonstrates the deficiency of brightness of the image for low tube voltages, caused by insufficient current delivery capacity of the generator. The "Iso-watt-point," that is, the tube voltage above which full power can be generated, is shifted to higher values than the tube alone would allow for.

3.11 Electron Emitter Heating and Emission Control

Cathodes of most diagnostic X-ray tubes are operated at negative potential of either half the tube voltage in bi-polar settings or full tube voltage. (Only some mammography units operate the cathode at ground potential.) Therefore, cathode control voltages for switching and electrostatic deflection, see below, and emitter heating power have to be generated at a negative high voltage level. Returning to Figure 3.5, the right picture shows emitter heating transformers which transform AC from ground to high-voltage potential. If compatible X-ray tubes comprise

![Figure 3.13](image-url)
more than one filament, this unit has typically multiple (e.g., two) independent sets of solenoids, as shown. Other than in the distant past, the heating circuitry is typically current controlled to eliminate effects of contact resistance in the high-voltage plugs and in the tube, wire heating, and so on, and to allow for various cable lengths without the need to re-adjust the heating voltage. However, fatigue of the cathode filament causes an increase of the wire resistivity, see Behling (2016). Given a fixed heating current, the power dissipated rises over time (and with it the rate of evaporation). As a result, the tube current rises for a fixed heating current. The generator control electronics react and reduce the pre-programmed heating current per technique factor setting. Closed loop control steers the filament heating current according to the desired tube current at a given tube voltage. During adaptation, the generator is automatically programmed to properly control a newly mounted tube with its individual emitter characteristics, see Figure 2.15. Depending on the individual tolerance situation, it may be necessary to re-adapt.

As a consequence of the characteristics of the metal and the high temperatures of the thermionic emitter during operation, the resistivity of the usual thermionic tungsten emitters is much lower at ambient temperature than during operation. Heating from ambient would take a relatively long time for the preparation of an exposure. Therefore, emitters are pre-heated, with some amperes, to temperatures slightly below the onset of relevant electron emission when the system is in a stand-by state. This pre-heating also reduces other thermo-mechanical distortion, for example of the focusing geometry, from changing cathode temperature.

3.12 Grid and Electrostatic Deflection Supply

The spectral integrity of the X-ray beam in pulsed operation may be impaired when the generator pulses the X-ray output by switching the high voltage on and off. For CT, cables are typically between 1 and 3 meters long, the tube is on for an entire exposure of several seconds duration, and tube currents amount to hundreds of milliamperes. Thus, ramping up and down the cable charge is a relatively simple task. The X-ray spectrum is reasonably well defined, as it should be to guarantee a reconstructed image free of artifacts. However, the time required for charging and discharging of up to 40 meters long high-voltage cables and smoothing capacitances in angiography systems may be of the order of milliseconds, in particular when the tube current is low for dose saving, for example for fluoroscopy. Transition times may be of the same order as the shortest pulse lengths, as Figure 3.14 demonstrates. For a long period of transition the tube voltage remains below the required level while the tube current is still high and the tube delivers X-rays to the patient. As the patient is getting more and more opaque the lower the tube voltage, the relationship between skin dose and detector signal worsens dramatically.

It is, thus, advised for dose reasons to reduce the tube current instead of the tube voltage. The tube voltage and, with it, the X-ray spectrum will then remain untouched during the entire switching event. Unwanted patient dose does not appear. Indeed, can cathodes with limited current delivery capacity be switched through biasing of the thermionic electron emitter with respect to the focusing electrode in which it is embedded? A negative control voltage of a few kilovolts is typically applied to cut off the electron emission. A grid switch supply unit may be placed close to the tube inside the tube housing. This solution is preferred by Philips for high-performance tubes for interventional angiography. The clean switching pattern is shown in Figure 3.15. Other

![Figure 3.14](image1.png)

**FIGURE 3.14** Disadvantage of pulsing the X-ray output for interventional fluoroscopic application by switching the tube voltage only. Compared with the desired pulse length, a long tail of production of soft X-rays during the decay of the high voltage drop appears. In this stage, the tube is still conductive and produces unwanted soft X-rays which enhance the skin dose of the patient without creating sufficient image data at the detector. Grid switching the tube current only solves this problem, see Figure 3.15, but requires additional electronics and tube features.

![Figure 3.15](image2.png)

**FIGURE 3.15** X-ray pulsing for interventional angiography with a grid switch. The tube current is switched using a bias voltage in the cathode. The tube voltage stays constant. The tube delivers an optimal X-ray spectrum. Dose savings in comparison with voltage pulsing shown in Figure 3.14 may amount to up to 20%.
3.13 Rotor Control

Rotor drives must match well with the connected tubes. Rotating frame tubes with their high-performance motors with exceptionally short air gaps pose other challenges compared to bi-polar rotating anode tubes, with air gaps of the order of 1 centimeter across vacuum insulation and a tube frame made of glass, ceramics, or highly resistive metal. In an attempt to shorten start-up time, bi-polar tubes have been constructed which employ grounded rotors and rotating insulators. Air gap and magnetic reluctance are, thus, minimized. The rotor control has to match the different conditions. Otherwise, depending on the moment of inertia of the anode, rotor inductance, bearing friction, and stator inductance, an improperly designed rotor control may cause excessive heating of the tube assembly. Other failure modes include insufficient final rotor speed in cases where the bearing friction is higher than expected. This may, in particular, happen for ball bearings with spring-thrust pre-load to avoid position dependent changes of the noise pattern or for tubes with large imbalance of the anode and intrinsic resonance below the nominal rotor speed. Improper driving momentum may result in “trapping” of the rotor close to the resonance, when the bearing friction is maximal due to anode vibration.

Whilst anodes with ball bearing systems often feature coast times of an hour and need active braking by feeding DC current into the stator to save bearing life, the hydrodynamic friction of spiral groove bearings is several orders of magnitude higher. Mechanical friction losses may reach several hundred watts, depending on the stiffness of the bearing and its nominal rotor speed. Rotor drives for these tubes with liquid bearings have to deliver up to 1 kilowatt of electrical power. Their ramp-up characteristics are of less importance than for tubes with ball bearings. However, during the initial start-up, sticking friction of the resting bearing has to be securely overcome. Continuous rotation at full speed has to be maintained with minimized power.

Usually, three-phase drives are being used. More details can be found in Behling (2016), Chapter 6.2.3.1.

3.14 Magnetic Focusing

In addition to tube voltage supply, delivery of tube and electron emitter heating current, and rotor drive, the task setting for modern generators has been extended to provide electrical energy for other loads, such as magnetic focusing and deflection systems. Particular challenges are the speed of deflection, electromagnetic compatibility, compactness, and energy efficiency.

Magnetic focusing and beam deflection were introduced on a broad scale in 2003 by Siemens for the Straton® tube. The technology was later extended to double quadrupole focusing by Philips for the iMRC CT tube in 2007 and, again, by Siemens in 2013 for the Vectron® tube. Active control of the focusing parameters is necessary for these magnetic systems. Inductive load and eddy-currents induced in the material of the tube frame limit the deflection speed. Thus, the maximal voltage output which such a unit can deliver is of particular importance.

The vast majority of medical X-ray tubes, however, are focused electrostatically by properly formed electrodes which surround the thermionic electron emitter in the cathode. If the emitter and focusing electrode are on the same potential and as far as space charge effects can be ignored, the electron beam path is independent of tube voltage and no active control is necessary. This changes when biasing is applied for electrostatic deflection or control of the focal spot size, see Behling (2016), Chapter 6.2.1.8.

3.15 Tube Temperature Supervision

Another task is safeguarding the integrity of the tube. As X-ray tubes usually do not comprise sophisticated intelligence, thermal modeling and thermal supervision are either allocated to the generator or to the X-ray system. Typically, multi-level finite element algorithms are implemented in most generators. These models predict temperatures for the end of the programmed application protocols and indicate potential thermal overload. Hazards from overheating the tube assembly are prevented by thermal switches or pressure switches at the tube housing, which feed fail-safe generator circuitry.

3.16 Dose Control

Interventional application requires automatic adjustment of the X-ray intensity, depending on the opacity of the volume to be X-rayed. Either the X-ray system controller or the generator may take the task of stabilizing the brightness of the image, without losing contrast. The sample generator depicted in Figure 3.2 has this function included. The software development requires substantial effort and expertise. Dose signals are delivered from the image receptor, a flat panel detector, a vacuum image intensifier, or an X-ray transparent ion chamber. Optimal contrast with minimal patient dose is achieved by starting with a relatively low tube voltage and a pre-defined tube current. Later, the generator adjusts the technique factors according to a pre-defined strategy. Motion blur in cardiac application is avoided, as the pulse length is kept under a critical level, for example 10 milliseconds.
The tube voltage is only raised to avoid photon starvation in the case when tube current and pulse length have hit programmed maxima. Aspects of tube life, control speed, image impression, detector performance, and so on, have to be considered.

### 3.17 Monoblocks

When limited instantaneous power and energy per patient are sufficient for the intended application, the high-voltage transformer, rectifier, and smoothing means may be small enough to place it into a single housing in combination with the X-ray tube. High-voltage cables and separate casings then become obsolete. Figure 3.17 depicts the interior of a so-called monoblock for an exemplary mobile C-arm system. These devices are available with a stationary as well as a rotating anode tube. For the exemplary unit, AC current with a moderate frequency of several kilohertz is supplied from an external power inverter. The X-ray tube is located in the metallic shielding cylinder close to the cover-lid of the device, shown at the bottom of the picture. Cylindrical coils left and right of the center constitute the positive (left) and negative (right) high-voltage transformer. Rectifier diodes and smoothing capacitors are covered with black foam, which bridges the entire unit from left to right. The stator coil is visible to the left of the tube, and one of the two filament supply transformers, one for each filament, is at the bottom right.

Monoblocks can also be found in legacy low-tier CT systems, mammography systems, and mobile general purpose X-ray equipment. Peak power rating and the maximal continuous heat dissipation are limited due to restricted space and surface area for cooling. The limited continuous heat dissipation in systems for surgery application may be further reduced by the blankets used to cover the unit and by hygienic considerations which would not allow the use of fans.

### REFERENCES


