PERFORMANCE AND PITFALLS OF DIAGNOSTIC X-RAY SOURCES: AN OVERVIEW

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Abstract— **Performance and reliability of medical X-ray sources for imaging humans are crucial from ethical, clinical and economic perspectives. This overview will treat the aspects to consider for investment of equipment for medical X-ray imaging. Recent X-ray tubes deliver enhanced reliability and unprecedented performance. But, metric for benchmarking has to change. Modern terms for product comparison still need to penetrate the market and to be implemented in practice. It is time to abandon *heat units* and comply with latest standards, which consider current technology.**

In view of the increasing number of interventional procedures and the risks associated with ionizing radiation, toxic contrast agents, and the potentially hazardous combination thereof, system reliability is of paramount importance. This paper will discuss tube life time aspects.

The speed of scientific and industrial development of new diagnostic and therapeutic X-ray sources remains high. Still, system developers and clinicians suffer from gaps between aspirations and reality in day-to-day diagnostic routine. X-ray sources are still limiting cutting-edge medical procedures. Undesired side-effects, wear and tear, limitations of the clinical work flow, costs, and undesired characteristics of the X-ray source must be further addressed. New applications and modalities, like spectral CT, and phase-contrast or dark-field imaging will impact the course of new developments of X-ray sources. High performance and flawless operation of the very special kind of vacuum electronics of X-ray tubes can only be safeguarded by quality manufacturing and highly skilled craftsmanship. Joint development of semiconductor and vacuum based electronics, i.e. X-ray generators and tubes, has proven key to success in the medical industry to guarantee a seamless match of the once separate devices. Thus, the terminology is expected to morph from X-ray *tube* and *high-voltage generator* to *X-ray source segment*.

Keywords— **X-ray tube, X-ray generator, tube failures, X-ray tube metric, reliability.**

XV. INTRODUCTION

Physics, history, technology and manufacturing of X-ray sources for medical diagnostic imaging are treated in many textbooks of medical imaging, usually as sub-chapters. A comprehensive description of the many aspects related to diagnostic X-ray sources, issues of reliability and their possible future can be found in [1]. Instead, the present paper will briefly touch a small selected sample of deficits and pitfalls, which are relevant for the practical clinical use.

XVI. HISTORY

The discovery of Conrad Wilhelm Roentgen, who created X-ray photons for the first time on purpose, which are capable of gathering detailed information about the anatomy of patients, was a quantum leap for medical diagnostics and therapy. Besides many laboratory type installations, the world's first X-ray machine in a clinic was installed and used in March 1896 in a patient preparation room of an operation theater in the New General Hospital (now UKE) in Hamburg, Germany. This city has since then been home of an industrial center of X-ray technology. Only a few places worldwide house the required expertise. Initially, the technology evolved with overwhelming enthusiasm and speed. But, at the time the hazards of ionizing radiation were unknown, yet. Highly flexible construction materials like glass provided lots of freedom of experimenting in those days.

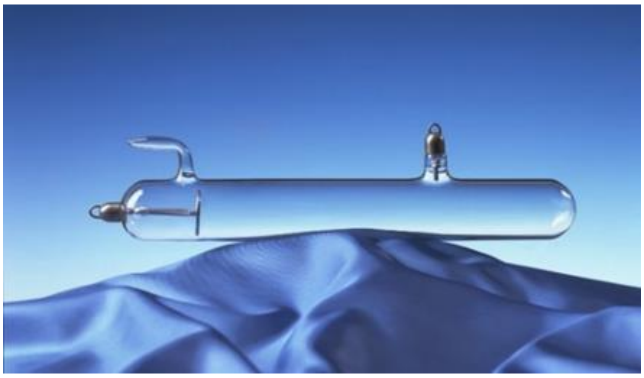

Fig. 1 Replica of an early ion X-ray tube as used by C. W. Roentgen. When the electron collector (short wire, right) is charged positively during the cycle of alternating high-voltage from the inductor and acts as an anode ions generated in the residual gas hit the aluminum slab on the left, the cathode. Cathode rays (electrons) are released, accelerated in the electric field, by-pass the anode wire and hit the glass wall at the right. Greenish fluorescence signals electronic current flow and X-ray generation. Backscattered electrons are collected by the anode wire. Roentgen wrote on Feb. 26th, 1896: "Namely, I discovered, that X-rays are not only created in glass, but also all other solid bodies (and perhaps even in liquids and gasses)…" (Translated from [2], pg. 56). (Photo curtesy of Philips.)

This extraordinary situation resulted in product cycle times of weeks rather than decades, which we are facing today. But, at the same time the extreme innovation rate was also paired with deep frustration. Summarizing an exhaustive year of ups and downs, Roentgen complained in January 1897 in a letter to his friend and assistant Ludwig Zehnder, see [2], pg. 65 (translated): "*Meanwhile, I have provisionally sworn, that I do not want to deal with the behavior of the [X-ray] tubes, as these dingus are even more capricious and erratic than the women.*" (Remark: A bowl of chocolate on his desk proves that Roentgen dearly loved his wife Bertha and vice versa, other than the above quote could suggest. To keep him happy, she replenished the chocolate daily. The bowl is still on exhibit at the German Roentgen Museum, Remscheid-Lennep, Germany.) But still, even after decades of technology development, the world of diagnostic X-ray sources surprises users and developers with pitfalls and in extreme cases, frustration. Besides evolving X-ray system technology, this is one of the many reasons for ongoing annual multi-million dollar industrial R&D investment.

XVII. X-RAY SOURCES

Roentgen's interface between tube and high voltage generator consisted of merely two wires: one charging the cathode, the other the anode, see Fig. 1. The tube current was simply defined by gas content and electrode geometry and was a simple monotonic function of the applied tube voltage. X-ray tubes and high voltage generators were developed in separate departments, sometimes even in separate companies. The first X-ray tube for use in a clinic came from C.H.F Mueller, later Philips, Hamburg, the inductor from Berlin, Germany. Today, experts prefer using the term *X-ray segment* instead of *tube* and *high voltage generator*. The complexity of the interface and the degree of interleaved R&D activity have both risen dramatically over time, latest with the advent of magnetic focusing and focal spot deflection. Other than with electrostatic focusing, magnetic control requires permanent fine tuning of the electrical supply of focusing magnets by the generator. Electron trajectories depend on the selected technique factors tube voltage and tube current which may vary during exposure. In addition, grid switching the tube current and focal spot deflection may require control voltages of several kilovolts to be supplied by the

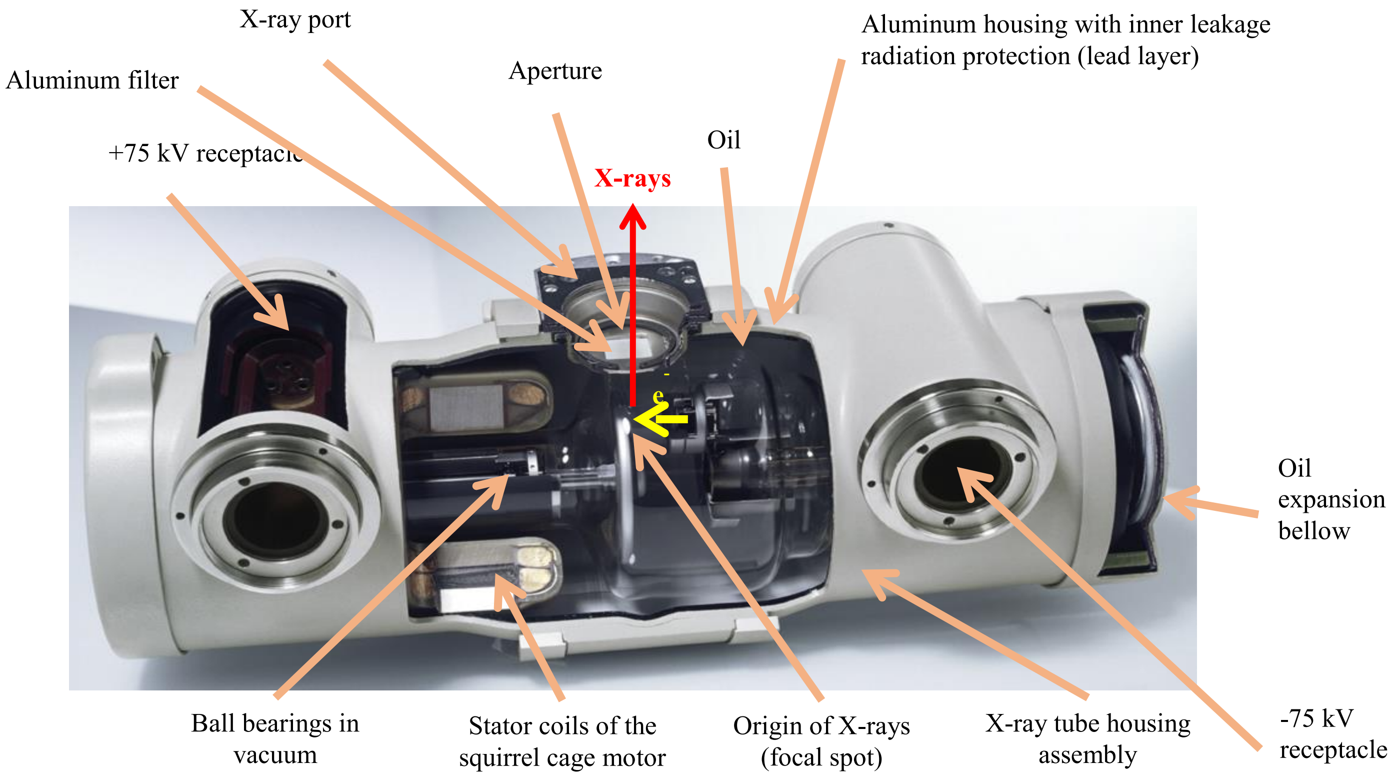

Fig. 2 Rotating anode X-ray tube housing assembly

generator to the cathode. Thermal and vacuum discharge management have become state-of-the-art. Intelligent X-ray dose control, rotor drives, safety management, service functionality, web connectivity add. The majority of large industrial vendors have therefore established integrated R&D departments, which have been given the joint responsibility for tube and generator development under single management. High performance X-ray tubes have morphed to X-ray source sub-segments of imaging systems rather than stand-alone items.

XVIII. BASICS OF FUNCTIONING

Before branching into the various possible pitfalls, it may be helpful to recall the basic functioning of rotating anode X-ray tubes. Figure 2 shows a sample cut model of a rotating anode X-ray tube housing assembly equipped with a rotor system with ball bearings, driven by an asynchronous squirrel-cage type motor. An electron beam is released into vacuum by a directly current heated thermionic tungsten electron emitter. Electrons are then accelerated in the electric field between cathode and anode. Photons emerge from a micrometer thin layer below the surface of the target during interaction of the electron beam with high-z and high density material, preferably tungsten. A small percentage of primary electrons shot into the extremely high electric field around the nuclei of the target convert their kinetic energy into electromagnetic X-radiation in the focal spot (FS). The tube voltage defines the X-ray photon spectrum with a Duane-Hunt cut-off energy at $-eU_t$, e being the electron charge, U_t the tube voltage. Undesired X-ray photons of lowest energy are taken out of the used beam. The soft end of the spectrum is defined by X-ray filters in the beam path, typically made of aluminum slabs with a thickness in the order of 2.5 mm or more, depending on the application. Due to their high absorption rate per length of tissue passed, photons with energies below ca. 15 keV to 30 keV would merely raise the patient's skin dose without delivering information to the detector.

The factor of conversion of electrical power to used X-ray intensity is in the order of 10^{-4}, for details see [1], chapter 2.2. Thus, typically the generation of bremsstrahlung (brake radiation) for imaging, means management of kilowatts of electric power and sophisticated heat management, see Figure 4.

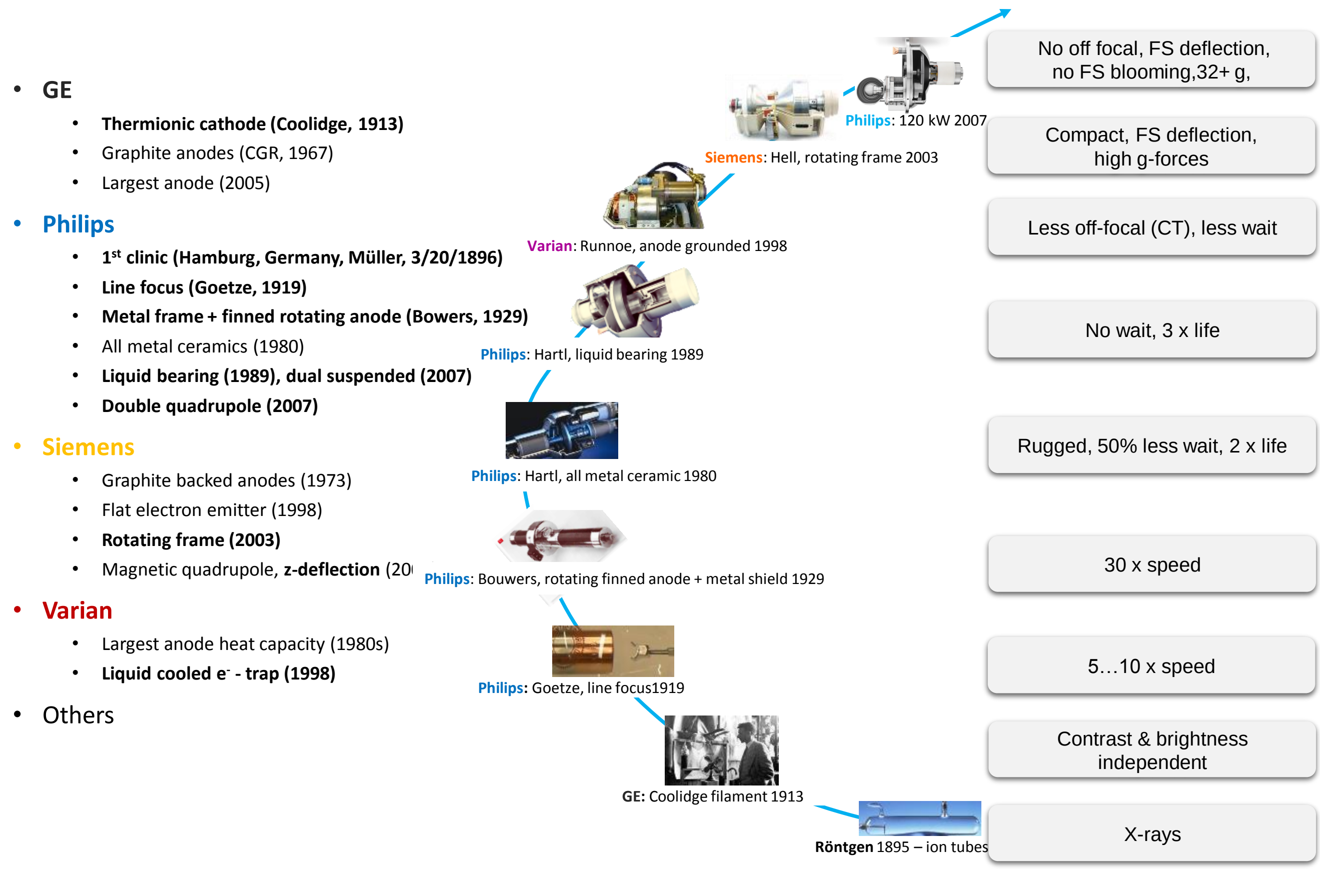

Fig. 3 Key achievements by various vendors (in alphabetic order) and their historic predecessors

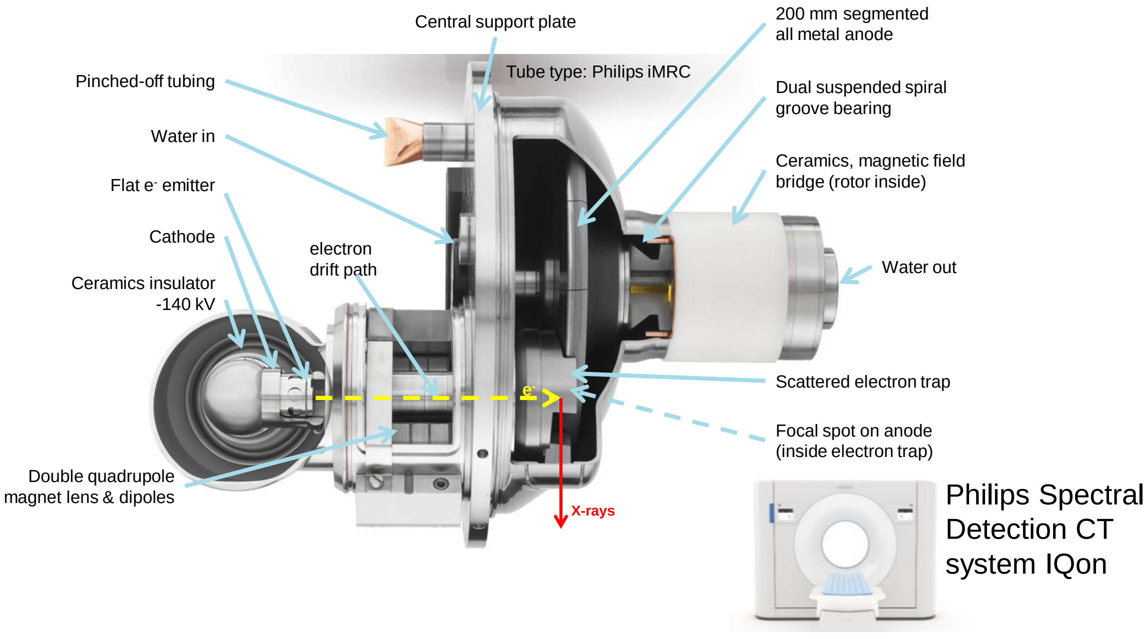

Fig. 4: Latest premium tube technology. Cut view of the Philips iMRC® tube in CT systems iCT ® and IQon®, see text. (Picture curtesy of Philips.)

XIX. IMPROVEMENTS

Roentgen's discovery started fierce race for technical improvement. Figure 3 points to major milestones. As a consequence of the great simplification of the way to operate X-ray systems, a division between medical physicists and radiologists occurred with Lilienfeld's and Coolidge's introduction of thermionic electron sources instead of a pure gas discharge. They both aimed at softening X-ray tubes. Lilienfeld attached an incandescent light bulb to the ion tube to boost electron production at low tube voltage. Coolidge, then working at GE, a little later inferred a ductile tungsten wire as the electron source and abandoned residual gas. From now on, contrast (spectrum, beam hardness) and brightness (tube current) of the X-ray image could be adjusted independently. Simplification of use allowed for a split of radiographers into medical physicists and radiologists. A significant improvement was the introduction of a rectangular Götze line focus by the manufacturer C.H.F. Müller, later Philips, which helped improving the brightness of the FS. This invention made use of the isotropic angular intensity distribution of bremsstrahlung which extends down to low take-off angles. The next quantum leap was accomplished by Bouwers of the Philips research laboratories, who industrialized a rotating anode with finned structure. Götze focus and rotating anodes helped cutting exposure times by more than two orders of magnitude. Siemens improved these anodes by speeding them up and backing the metal disk with graphite. Again, Philips resumed the lead and implemented all metal ceramics technology in medical tubes, which has become state-of-the art for the entire industry. Wait times could be reduced and eventually completely eliminated by introducing hydrodynamic bearings and unprecedented large anodes in computed tomography and interventional X-ray systems. Varian bettered the heat balance by improving the capturing of back-scattered electrons, dissipating their waste energy and reducing off-focal radiation. A quest for compactness and operation at high centrifugal forces stimulated Siemens to realize an idea from General Electric and Metropolitan Vickers of the late 1940ies: The rotating anode was made an integral part of the now rotating tube frame with immediate contact to the surrounding oil. The rotating frame tube Straton® was born in 2003 which allows for focal spot deflection in azimuthal and axial direction, see [3] and [1], chapter 1.3.9. Some limitations of FS power density and FS stability at high tube current became apparent, however.

With their iMRC platform Philips introduced the latest major technological leap. The company launched in 2007 the iMRC® tube, which is shown as a cut model in Figure 4. The iMRC® tube family is a premium tier CT tube platform which allows for high speed rotation of a segmented anode at high centrifugal acceleration in a fast revolving CT gantry. Unprecedented FS power density for very high spatial image resolution is accompanied by high and exactly shaped photon flux for low image noise. This is of particular importance for detection based spectral CT imaging. The electron beam of the iMRC® tube originates from a directly heated meandered surface of a comparatively large planar thermionic tungsten emitter. This technology combines mechanical and chemo-physical robustness and totally eliminates for the first time space charge limitations of the tube current in rotating anode tubes. Double quadrupole magnetic focusing and deflection and a highly efficient electron trap eliminate FS blooming, and artifacts from aliasing and off-focal radiation, see below. Siemens adapted their latest tube development to this set of technologies with their Vectron® tube and launched it in 2013. More details are in [1], chapter 1.3.12.

XX. Latent pitfalls

Despite of great progress, X-ray tube technology still suffers from a number of technical pitfalls, which should be known to assess and avoid image artifacts, optimize system design and enable efficient fault finding in practice. Attention should be paid to tube life time which may depend on gantry speed in CT, gyroscopic forces in interventional systems, and power requested. Other aspects comprise degradation of the dose output over time, preparation time, electrical stability and means in the generator to cope with vacuum discharges. Image quality and spatial definition depend on focal spot size and stability under rotation, thermal capacity, off-focal and leakage radiation, and scattered radiation from the tube port. The work flow depends on cooling time, short and long term.

XXI. Thermal issues and new metric

The inefficiency of X-ray generation has produced the most severe bottleneck for the clinical routine: target cooling, see figure 5. Stationary anodes were the only available technology for more than three decades, which resulted in long exposure times and motion blur. The focal spot of Bouwers' rotating anode is cooled by convection cooling, instead. Heated material is removed by rotation. Cooling of the bulk material is challenging, however. Ball bearings, which were initially employed and which have become industry standard until now for conventional tubes, substantially block heat conduction. The alternative thermal radiation is efficient at high temperatures, but ceases with fading visible glow of the body. Thus, heat had to be stored in the target before heat radiation could slowly dissipate it in preparation of the next patient. In view of this technology, anode heat storage capacity has become the primary figure of merit for tube selection from the 1930ies. More *heat units* (HU) suggested better performance. The terms *heat unit* or *mega heat unit* (MHU) have never been exactly standardized, and originally refer to rotating anode X-ray tubes supplied by dual-phase high voltage generators, an electronics technology of the 1920ies.

The IEC began specifying the storable heat content in SI units instead, defined single level heat integration algorithms in the form of heating and cooling charts and the method of validation of the heat storage capacity of an anode. Eventually, by the launch of novel tube and generator technology *maximum anode heat content* turned from a historic key performance indicator into a confusing term. As a consequence of the advent of electron traps in metal ceramics tubes (Philips SRC, molybdenum aperture, 1980), heat conducting liquid metal hydro-dynamic bearings (Philips MRC, 1989), and definitely by the launch of rotating frame technology (Siemens Straton®, 2003, see [3]), the IEC amended the standard IEC 60613 in 2010. Without even mentioning anode heat content anymore, tube performance, e.g. for CT, is now simply rated by the *Nominal CT power*, defined as the power which a tube can sustain during a demanding realistic sequence of scans: 4 seconds exposure within an endlessly repeated cycle of 600 s duration each. Other than before, compliance with this metric can most simply be validated by the user. It is time to finally abandon HU's and MHU's for tube comparison.

XXII. Current issues

There are other pitfalls related to the production of electrons and their fate after impact on the target. Figure 6 shows an exemplary electron emission curve for an interventional angiography tube with tungsten coil electron emitter. The characteristics is best described by both, space charge limited emission according the Child-Langmuir $V_t^{3/2}/d^2$ - law for high currents, V_t being the tube voltage and d the cathode-to-anode distance, and the Richardson equation with its exponential temperature dependency, which is appropriate for low currents.

The left part of the chart, where the tube current is small and the emitter temperature low, is dominated by thermionic emission without major space charge effects. The emission current density j_c and the tube current I_{tube} both rise steeply with temperature according to Richardson's law $j_c \propto T^2 e^{-Ww/(k_B/T)}$, where T is the temperature of the tungsten emitter with its work function W_w and k_B is Boltzmann's constant. For the sample tube shown the temperature T is rising about linearly with the indicated heating current I_{fil}. The majority of emitted electrons successfully escape from the cathode and are collected by the anode. Tube voltage does not matter in this saturation emission approximation (left). Ideally, the tube current should be unlimited in the entire specified range of tube voltages and tube power. Generally, this is indeed so for stationary anode tubes. But, the permitted

Fig. 5 Thermal picture of a rotating anode in operation. The rectangular focal spot, 10 mm in length and about 1 mm wide, points radially and is located at the right on the anode, trailed by a comet tail of the hot surface of the focal spot track. The X-ray focal spot is located about half a focal spot width further to the right. Some reflected light is visible from the heated electron emitter coil in the cathode at the left. The anode rotated counter clock wise with about 50 Hz. (Adapted from [1].)

current and power density in the focal spot of a rotating anode is by one to two orders of magnitude higher. Negative electronic space charge between cathode and anode limits tube performance at low tube voltages and high tube currents. Indicated by the blue squares in Figure 6 at the maximum permitted temperature, coiled electron emitters present an approximately linear decrease of the

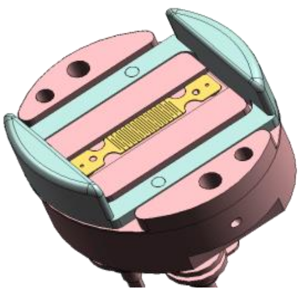

Fig. 7 CAD model of the cathode of the Philips iMRC® tube for CT (systems iCT® and IQon®). The use of a meandered flat electron emitter (yellow) enables generation of unprecedented high tube currents even at low tube voltages. The emitter is robust enough to withstand adverse conditions of residual gas pressure, ion bombardment and vacuum discharges. Space charge limitations are practically absent.

maximal tube current I_{tube} with the tube voltage in practice. Only for voltages above the so-called isowatt point, where cathode and anode performance meet, the anode is the limiting component, indicated by red circles.

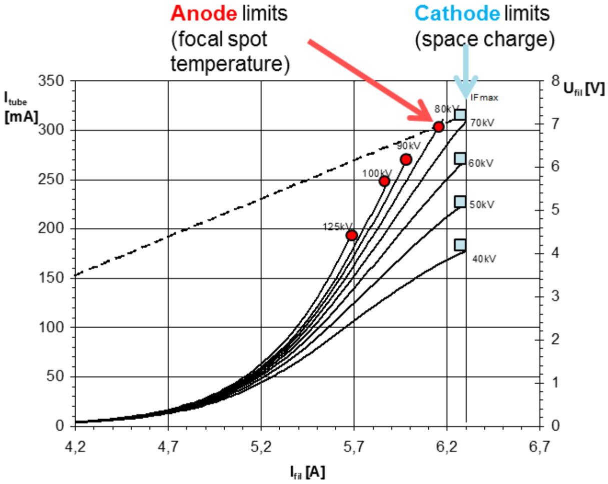

Fig. 6 Sample emission curves for the small focal spot of the rotating anode tube Philips MRC 0407 with tungsten coil emitters. I_{tube} (left axis) represents the tube current, which depends on tube voltage (parameter indicated) and heating current I_{fil} (abscissa). Red circles indicate anode limits, blue squares cathode limits. Dotted curve: U_{fil} (right axis) is the voltage to drive the filament heating current I_{fil}. (Graphics adapted from [1].)

The approximately linear relationship between maximal permitted tube current tube voltage at low electric field strengths results from a mixed type of emission, the contribution of thermionic and space charge limited electron emission from the different surfaces of the emitter. As said, the Child-Langmuir law would suggest proportionality between I_t and $V_t^{3/2}$ for pure space

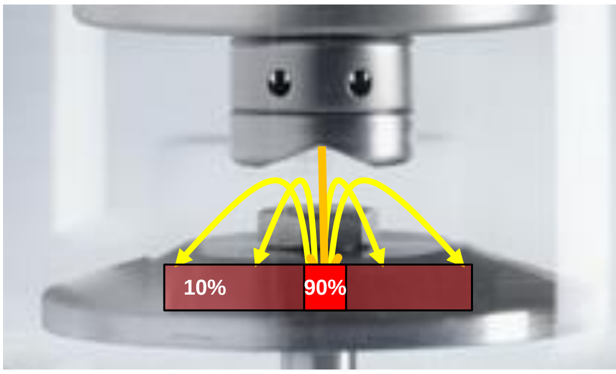

Fig. 8 Backscattered electrons (yellow) causing off-focal radiation in a bipolar glass tube. These electrons are reflected by the cathode (in this design facing the anode), hit the anode outside the border of the primary focal spot (red). Cathode: top, rotating anode: bottom. Percentage figures roughly quantify the intensity of X-rays. (Adapted from [1].)

charge limited emission at low tube voltages and high filament temperatures. But, for the exemplary cathode referenced in Figure 6 this behavior is somewhat obscured. Regions on the emitter which are subject to large enough electric fields, which suppress the space charge effects, add current and partly counter-balance the current deficit. The advent of flat emitters greatly improved the situation, see Figure 7. With reference to figure 6, the absence of space charge limitations would mean that all emission curves would coincide with the 125 kV-curve and extend beyond it to the right, no matter how small the tube voltage would be within the interesting range.

XXIII. Impaired image quality and remedy

Another pitfall is caused by the intense interaction of electrons with the target material tungsten. The origin of X-rays in conventional glass tubes, as shown in Figure 2, is typically an extended area, about a few micrometers below the surface surrounding the center of the intersection of the beam of primary electrons with the anode surface. Scattering of electrons at the nuclei of the target creates photons, but also severely alters the direction of the electrons. Given the high electric potential inside the tungsten nuclei, which peaks at more than 20 megavolts, a few scattering events will suffice to cause rapid angular diffusion of the primary electrons in the vicinity of the surface. Thus, about half of the impinging particles are backscattered and lost into vacuum. On average, they carry nearly 40% of the primary power. In typical standard tube designs, where the cathode is located directly facing the anode, like the one shown in Figure 2, backscattered electrons will be mirrored back and experience a second impact. Soft off-focal radiation is generated outside the desired focal spot and may impair the definition of the image. It may cause shadows around highly contrasting objects, like iodinated vessels in angiography application.

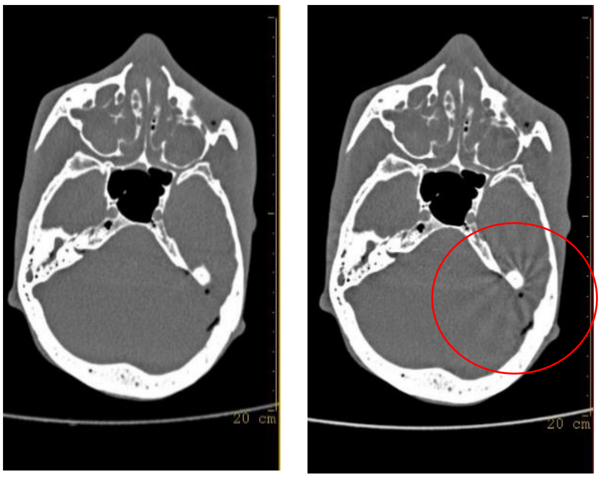

Fig. 9 Suppression of so-called windmill artifacts at sharply contrasting edges using axial deflection of the focal spot (FS) position between projections in CT. (Adapted from [1].)

Off-focal radiation may even mimic a bleeding and confound the differentiation between hemorrhagic and ischemic stroke. Therefore, electron traps in high performance tubes, as shown in figure 4, are built in to gather the majority of these backscattered electrons. For those tubes, like the one shown in Figure 4, X-ray scattering in the tube window remains as the only cause of off-focal radiation.

XXIV. Suppression of artifacts

Modern features of X-ray sources augment artifact suppression in computed tomography. Figure 9 demonstrates the efficient remediation of aliasing artifacts by axial and azimuthal deflection of the FS between projections in CT. The electron beam of the suitable X-ray tube, as shown in figure 4, is magnetically deflected in radial direction by about half a focal spot length, which translates to an apparent axial displacement of the projected focal spot. The additional information can be used by the reconstruction algorithm to suppress so-called windmill artifacts at sharp edges in axial direction, as shown in figure 9.

XXV. Consumable X-ray tube

The most costly issue with medical X-ray tubes remains short tube life. There is little public data available. Erdi, see [4] and Table 1, reports an average of 19.2 and 22.4 months in 50 replacements of high-performance CT tubes at 13 GE scanners in the Sloan Kettering Center NY, USA. But, tube life differs significantly between vendors. Differences of settings in imaging systems, local adaptations and preferences, frequency and ways of usage further broaden the life time distribution. Outliers are often caused by defects in the rotor drive, long periods of operation in preparation mode, extraordinary high tube voltages or unusual low tube voltages and high emission currents used. Other than e.g. for incandescent light bulbs, the temporal failure distribution for X-ray tubes is typically very broad, as can be seen from the columns Spread and Min-max. in Table 1. This underscores the need of commercial insurance policies, ideally by signing tube-included service contracts with the supplier. There are many ways for the user to save costs and extend reliability. Typical causes of failure and Pareto distributions are treated in [1], chapter 9.5.

Table 1 Average Tube life of GE CT tubes (adapted from [4], see text).

Tube type	Av. Life (months)	Spread (months)	Min.-max. (months)
GE Performix Ultra ®	19.2	±12.5	7-48
GE Performix Pro ®	22.4	±9.6	12-32

The multitude of typical failure modes, e.g. for typical CT tubes, includes arcing, vacuum leakage and subsequent vacuum discharges, see figure 10, run-away arcing by thermal overheat of the target, electron field emission from irregularities on electrodes and loose particles, low X-output, hardening of the X-ray beam, pollution of the X-ray window, notably by carbonization of cooling oil, rotor vibration, bearing noise, frozen rotors, evaporated electron emitters and short circuits, implosion of glass tubes, damaged or polluted heat exchangers and pumps, broken anodes, burnt-out stator coils, damaged mechanics, and more.

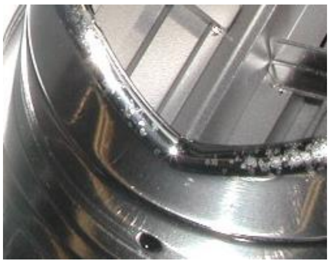

Fig. 10 Foot point craters on a used cathode head after severe vacuum discharge activity (arcing), adapted from [1].

XXVI. Conclusion

A number of pitfalls and limitations of current technology of generation of medical diagnostic X-rays still exist and demand attention by system developers, medical physicists and clinicians.

Instead of adhering to outdated terminology, selection and investment of medical diagnostic X-ray sources should be supported by applying the latest metric and standards which reflect most recent technology.

Tube life remains an issue. Commercial risks should be managed by selecting experienced suppliers which offer well matching premium quality tubes and generators. The commercial burden of tube failure may be imposed on the vendor by signing tube-included service contracts.

References

1. Behling, R., 2016. *Modern Diagnostic X-Ray Sources: Technology, Manufacturing, Reliability*. CRC Press, Taylor & Francis, Boca Raton, USA. ISBN 978-1-4822-4132-7.
2. Zehnder, L. 1935. *W. C. Röntgen – Briefe an L. Zehnder. (*W. C. Roentgen – letters to L. Zehnder). Switzerland: Rascher & Cie, AG Verlag.
3. Schardt, P. et al. 2004. New X-ray tube performance in computed tomography by introducing the rotating envelope tube technology. *Med. Phys.* 31(9): 2699–706.
4. Erdi, Y., E. 2013. Computed tomography X-ray tube life analysis: A multiyear study. *Radiologic Technology* 84: 567.

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