Chapter 5: X-Ray Production

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Diagnostic Radiology Physics:
A Handbook for Teachers and Students

Objective:

To familiarize the student with the principles of X ray production and the characterization of the radiation output of X ray tubes.

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Slide set prepared
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following initial work by
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5.1 INTRODUCTION

Basis of Radiological Imaging

The differential absorption of X rays in tissues and organs, due to their atomic composition

X-Ray Production

Principles have remained the same since their discovery however many design refinements have been introduced

This Chapter

Outlines the principles of X ray production and characterizes the radiation output of XRTs

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5.2 FUNDAMENTALS OF X-RAY PRODUCTION

The **Production of X Rays** involves the bombardment of a thick target with energetic electrons

Electrons undergo a complex sequence of collisions and scattering processes during the slowing down process which results in the production of

- **Bremsstrahlung** and
- **Characteristic Radiation**

A **Simplified** treatment of this process, based on classical theory, is provided in this section

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Energetic **Electrons** are mostly slowed down in matter by:

- Collisions and
- Excitation interactions

If an electron comes close to an atomic **Nucleus** the attractive Coulomb forces causes a change of the electron's trajectory

An accelerated electron or an electron changing its direction emits electromagnetic radiation and given the name

Bremsstrahlung

(braking radiation)

The energy of the emitted photon is subtracted from the kinetic energy of the electron

The **Energy** of the Bremsstrahlung photon depends on the

- **Attractive** Coulomb forces and hence on the
- **Distance** of the electron from the nucleus

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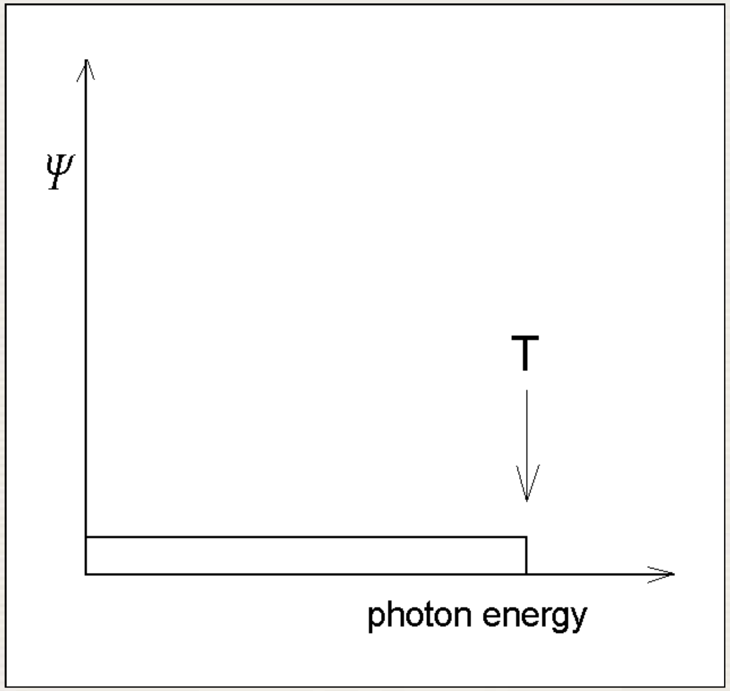
Classical Theory

Consider that electron bombardment of a thin target yields a constant energy fluence (Ψ) from zero up to the initial electron kinetic energy (T)



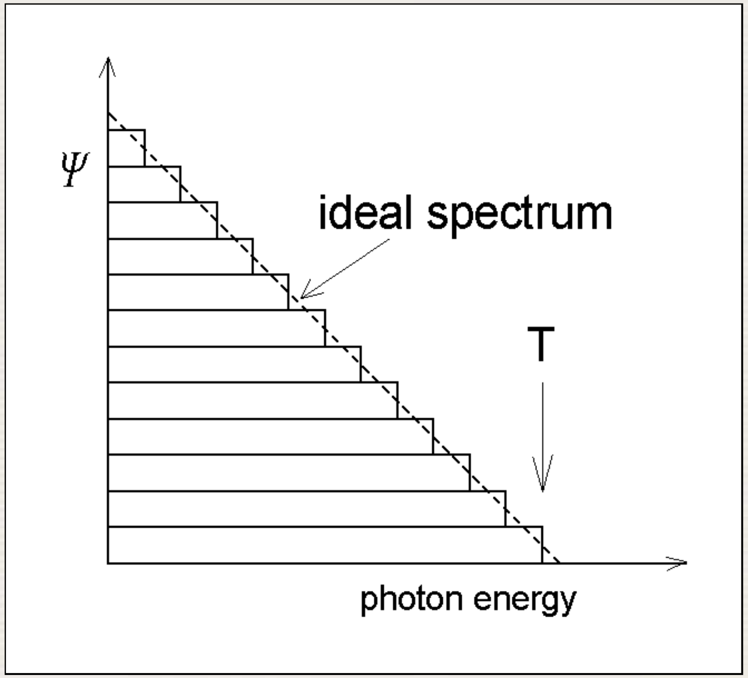
5.2 FUNDAMENTALS OF X-RAY PRODUCTION

5.2.1 Bremsstrahlung

A thick target can be thought of as a sandwich of many thin target layers each producing a rectangular distribution of energy fluence

The superposition of all these rectangular distributions forms a **triangular** energy fluence distribution for a thick target, the

Ideal Spectrum



The ideal spectrum does not include any attenuation effects

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According to this model

An increase in electron energy increases the number of thin layers each radiating X rays

The triangular area grows proportional to the square of the electron energy

Therefore, the **Radiation Output** of an XRT is proportional to $\mathbf{U^2}$

U: tube voltage

relationship holds if spectral changes due to attenuation and emission of characteristic radiation are ignored

A **Fast Electron** colliding with an electron of an atomic shell could knock out the electron once its KE exceeds the binding energy of the electron in that shell

The binding energy is **Highest** in the most inner K-shell and decreases for the outer shells (L, M, ..)

The **Scattered** primary electron carries away the difference of kinetic energy and binding energy

The vacancy in the shell is then filled with an electron from an outer shell accompanied by the emission of an **X Ray Photon** with an energy equivalent to the **Difference** in binding energies of the shells involved

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For each element binding energies and the **Monoenergetic** radiation resulting from such interactions, are unique and **Characteristic** for that element

Element	Binding energy, keV		Energies of characteristic X rays, keV			
	L-shell	K-shell	K_{a1}	K_{a2}	K_{b1}	K_{b2}
W	12.10/11.54/10.21	69.53	59.32	57.98	67.24	69.07
Mo	2.87/2.63/2.52	20.00	17.48	17.37	19.61	19.97
Rh	3.41/3.15/3.00	23.22	20.22	20.07	22.72	23.17

Instead of characteristic radiation the energy available could be transferred to an electron which is ejected from the shell (**Auger Electron**) - production probability decreases with Z

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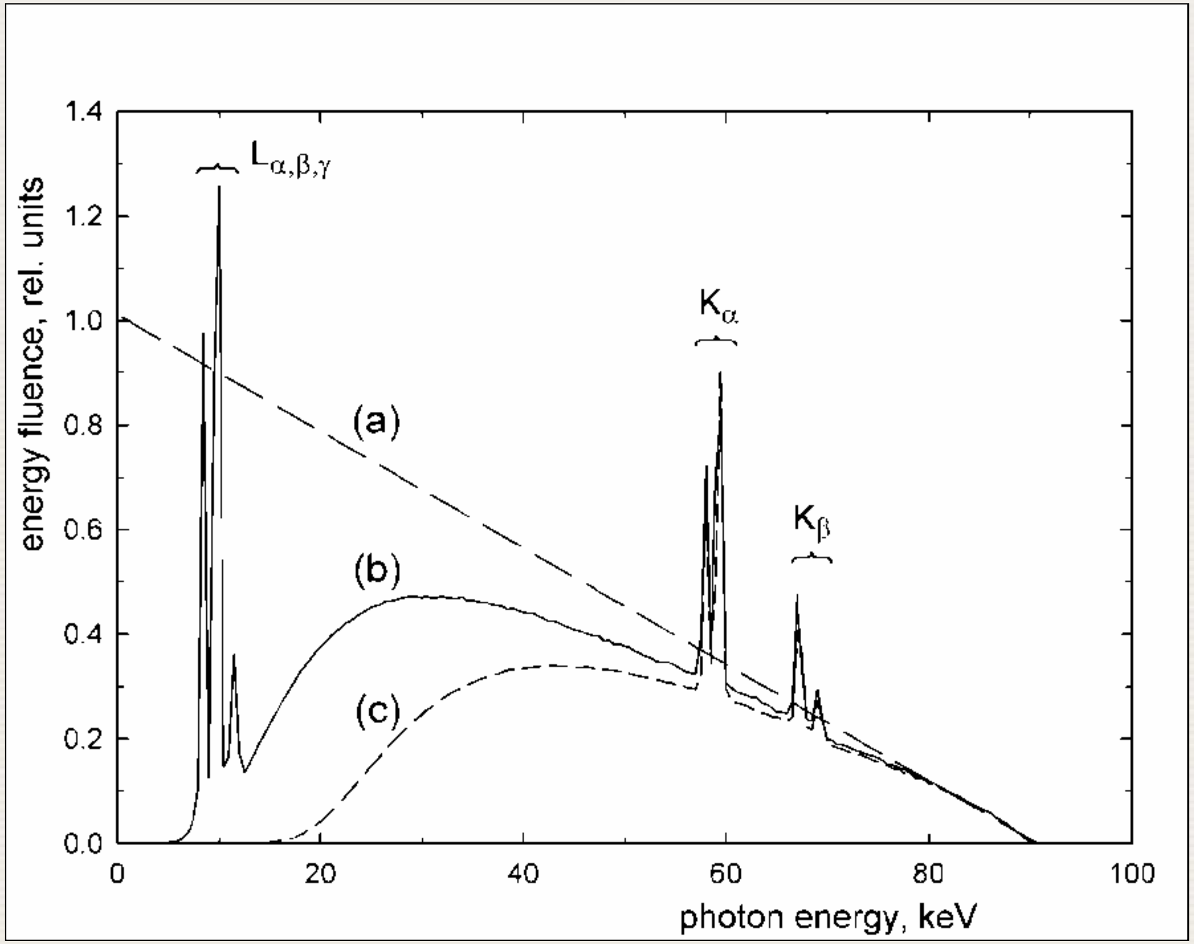
5.2 FUNDAMENTALS OF X-RAY PRODUCTION

5.2.3 X-ray Spectrum



a) Ideal **Bremsstrahlung** spectrum for a tungsten anode (tube voltage 90 kV)

b) An **Actual** spectrum at the beam exit port with characteristic X rays (anode angle: 20°, inherent filtration: 1 mm Be)

c) The spectrum **Filtered** with an equivalent of 2.5 mm Al

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5.2.3 X-ray Spectrum

The electrons are slowed down and stopped in the **Target**

within a range of a few tens of μm

X rays are not generated at the surface but within the target

resulting in **Attenuation** of the X ray beam

Self-Filtration appears most prominent at the low-energy end of the spectrum

Characteristic Radiation shows up if the kinetic energy of the electron exceeds the binding energies

L-Radiation is totally absorbed by a typical filtration of 2.5 mm Al

The **K-Edge** in the photon attenuation of tungsten can be noticed as a drop of the continuum at the binding energy of 69.5 keV

For tungsten targets the fraction of K-radiation contributing to the total energy fluence is **<10%** for 150 kV tube voltage

The **Radiative Mass Stopping Power** of electrons is proportional to $\mathbf{Z^2}$

Integration along the electron path gives the total X ray energy fluence as

$$\Psi \sim Z \cdot I \cdot U^2$$

where I: electron current and U: tube voltage

If a high Bremsstrahlung yield is required, metals with high **Z** are preferable

Tungsten (Z=74) is commonly chosen as it also withstands high temperatures (2757°C at $1.3 \cdot 10^{-2}$ Pa vapour pressure)

Efficiency for the conversion of electrical power to Bremsstrahlung radiation is proportional to **U·Z**

At 100 kV the efficiency is as low as **~0.8%**

This is the cause for most of the technical problems in the design of XRTs as practically all electrical power applied in the acceleration of electrons is converted to **Heat**

5.2.3 X-ray Spectrum

The ideal spectrum appears triangular with the **Energy Fluence** taken as the quantity describing the spectral intensity

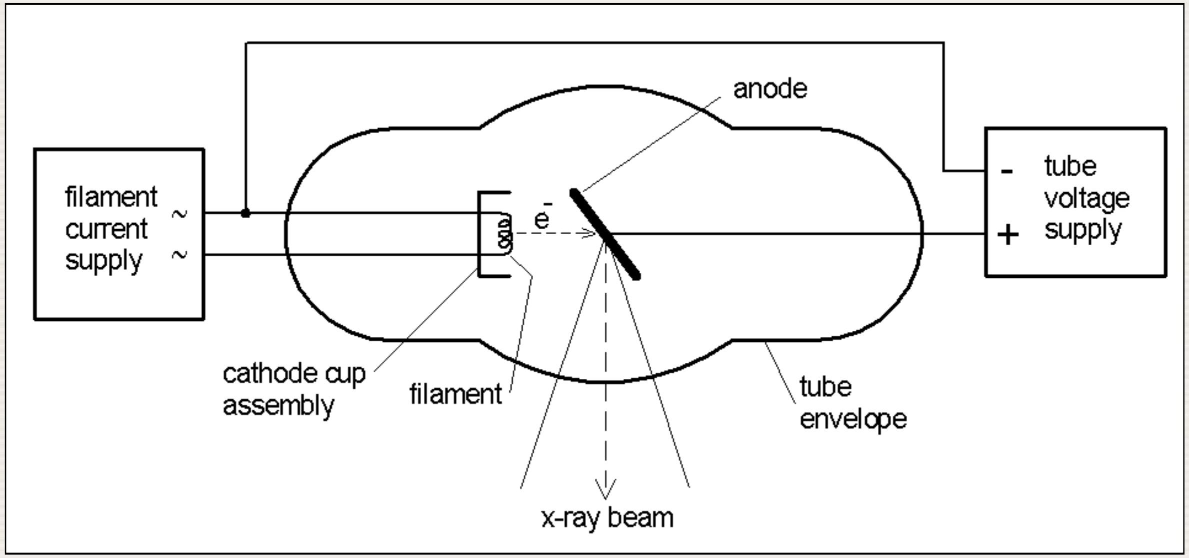
The **Photon Fluence** is a more practical quantity for calculations using spectral data and is therefore used in the following sections

More refined models for the generation of X ray spectra have been developed using **Monte Carlo methods**

For practical purposes a **Semi Empirical** approach gives satisfactory results, useful in simulations

5.3.1 Components of the X Ray Tube

The production of both **Bremsstrahlung** and **Characteristic Radiation** requires energetic electrons hitting a target



Principle components of an X ray tube are an **Electron Source** from a heated tungsten filament with a focusing cup serving as the tube **Cathode**, an **Anode** or **Target** and a **Tube Envelope** to maintain an interior vacuum

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5.3 X-RAY TUBES

5.3.1 Components of the X Ray Tube

The **Filament** is heated by a current that controls the thermionic emission of electrons, which in turn determines the number of electrons flowing from cathode to anode (**Tube** or **Anode Current**)

e.g. <10 mA in fluoroscopy and 100 to >1000 mA in single exposures

The accelerating **Potential Difference** applied between cathode and anode controls both X ray energy and yield

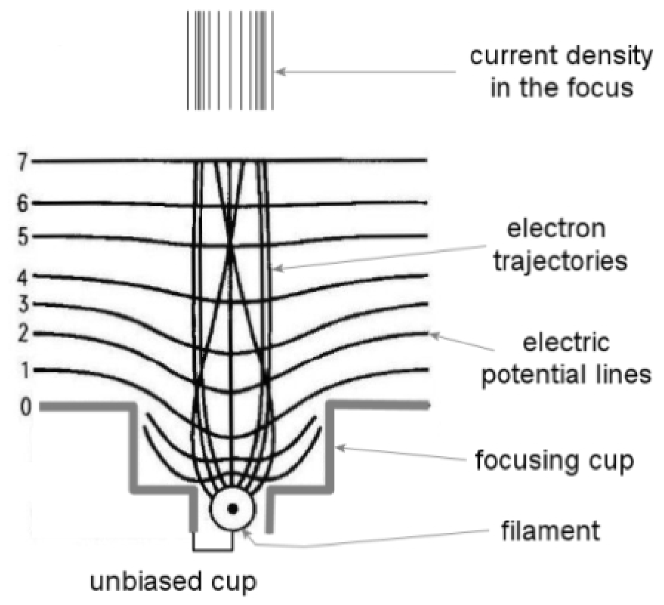
e.g. 40 to 150 kV for general diagnostic radiology and 25 to 40 kV in mammography

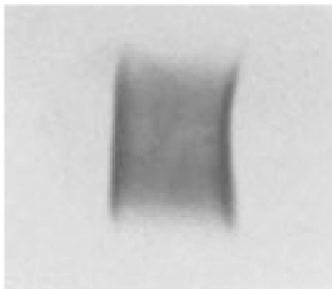
Thus **Two** main circuits operate within the XRT:

- Filament circuit
- Tube voltage circuit

The **Arrangement** of the filament, the focusing cup, the anode surface and the tube voltage generates an electric field accelerating the electrons towards the focal spot at the anode

The effect of an **Unbiased** focusing cup on the electric field and electron trajectories

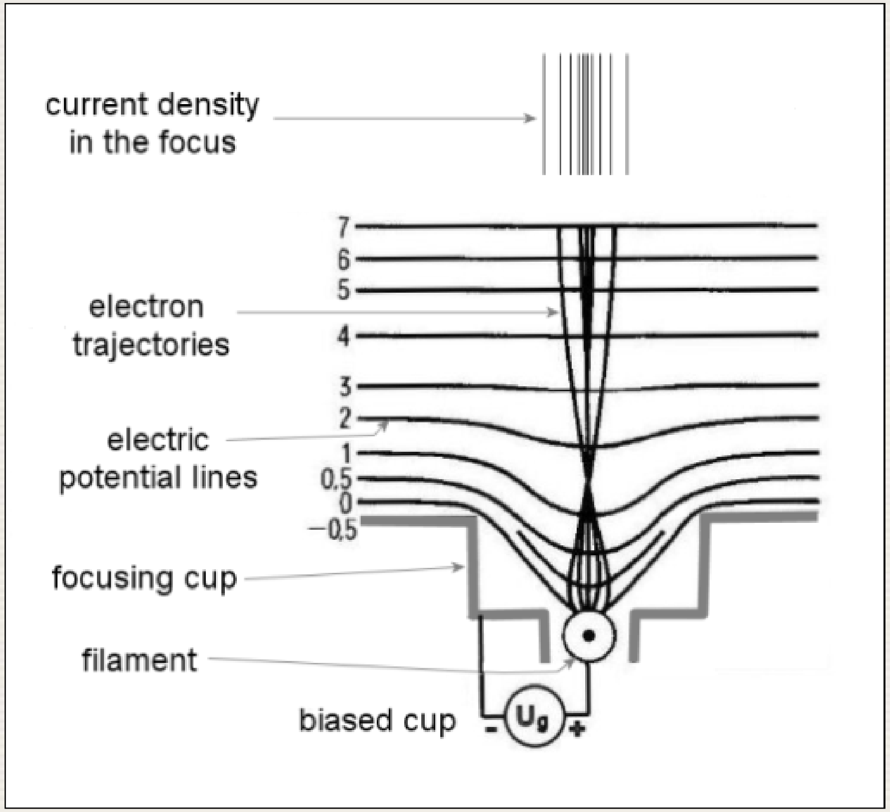




The typical **Bimodal** distribution of the current density can be seen in a pinhole image of the focus

Numbers indicate potential difference near the cup in kV

Biasing the focusing cup leads to a compression of the trajectories giving a smaller focus



Numbers indicate potential difference near the cup in kV

With an increasing negative bias voltage at the focusing cup the focus size will decrease and finally the electron current will be pinched **Off**

Effect is sometimes used to electronically control the focus size or for a fast switching of the anode current (**Grid Controlled Tubes**) when short radiation pulses are required as in pulsed fluoroscopy

The **Spiral-Wound** filament is typically made from tungsten wire of 0.2 to 0.3 mm diameter and operates at around 2700^{o} K

To minimise the **Evaporation** of tungsten from the hot surface, the filament temperature is kept at a lower level except during exposure when it is raised to operational levels

Thermionic Emission of electrons increases with temperature (Richardson's law) and produces a cloud of electrons (**Space Charge**) enclosing the filament

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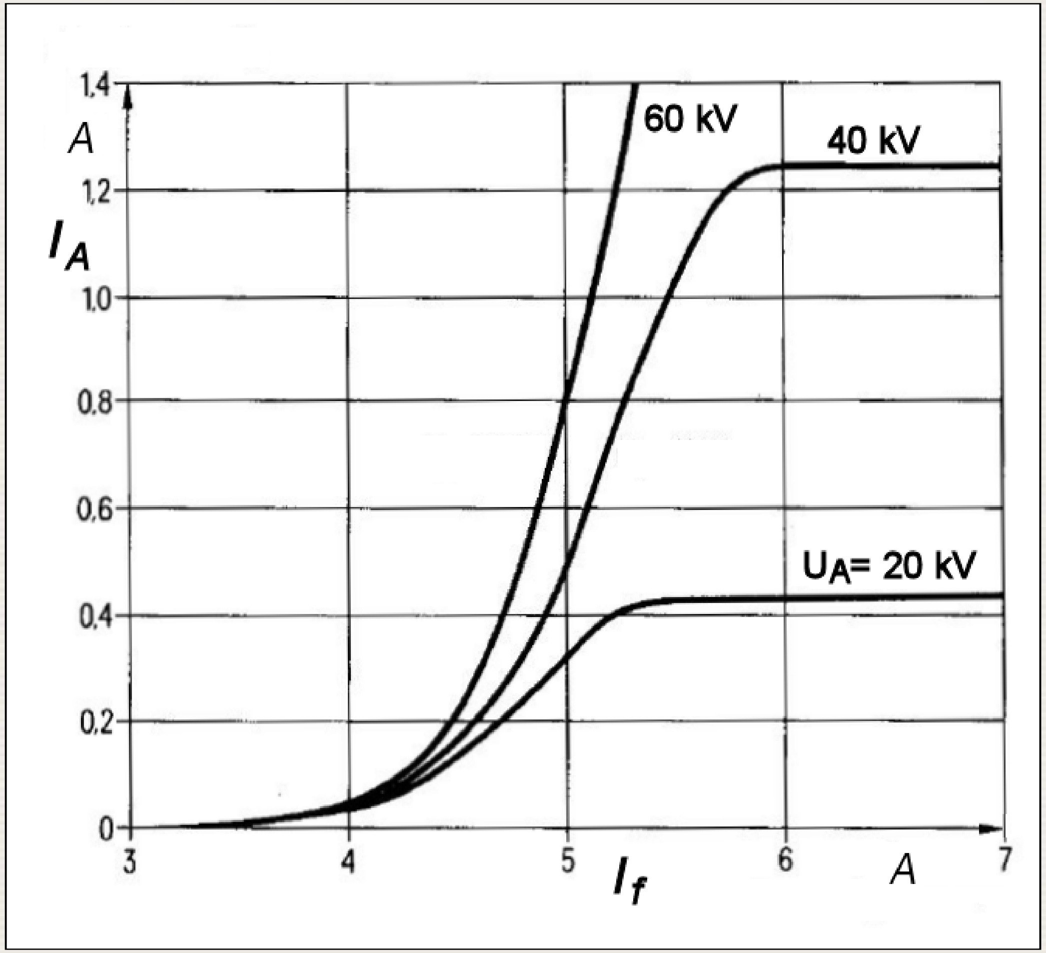
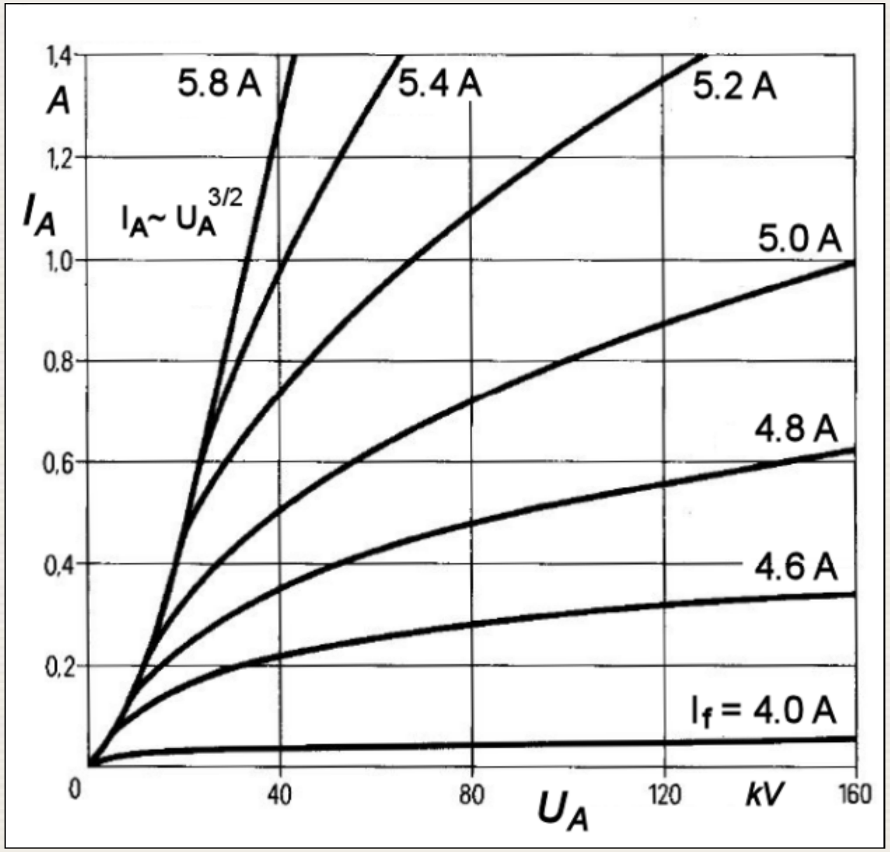
Space Charge shields the filament from the anode voltage

Anode current $\mathbf{I_A}$ vs. filament current $\mathbf{I_f}$ for various anode voltages $\mathbf{U_A}$ showing space-charge limited anode currents for the lower tube voltages



For **high** anode voltages all electrons boiled off the filament are accelerated to the anode giving an anode current fairly independent of tube voltage (saturation current)

Tube current **I_A** vs. tube voltage **U_A** depending on filament current **I_f**



Note: current saturation occurs for the lower filament currents

Choice of Material

For common radiographic applications a high **Bremsstrahlung** yield is mandatory requiring materials with high atomic numbers (**Z**)

Additionally, due to the low efficiency of X ray production it is also essential that the thermal properties such as **Maximum Useful Temperature** determined by melting point and vapour pressure, heat conduction, specific heat and density are also considered

Tungsten (Z=74) is the optimum choice

Choice of Material

For **Mammography** other anode materials such as molybdenum (Z=42) and rhodium (Z=45) are frequently used

For such anodes X ray spectra show less contribution by Bremsstrahlung but rather dominant **Characteristic** X rays of the anode materials

Allows a more satisfactory **Optimization** of image quality and patient dose

In **Digital Mammography** these advantages are less significant and some manufacturers prefer tungsten anodes

Line-Focus Principle (Anode Angle)

For measurement purposes the **Focal Spot Size** is defined along the central beam projection

For the sake of high anode currents the **Area** of the anode hit by the electrons should be as large as possible to keep power density within acceptable limits

To balance the need for large heat dissipation with that of a small focal spot size the **Line-Focus Principle** is used

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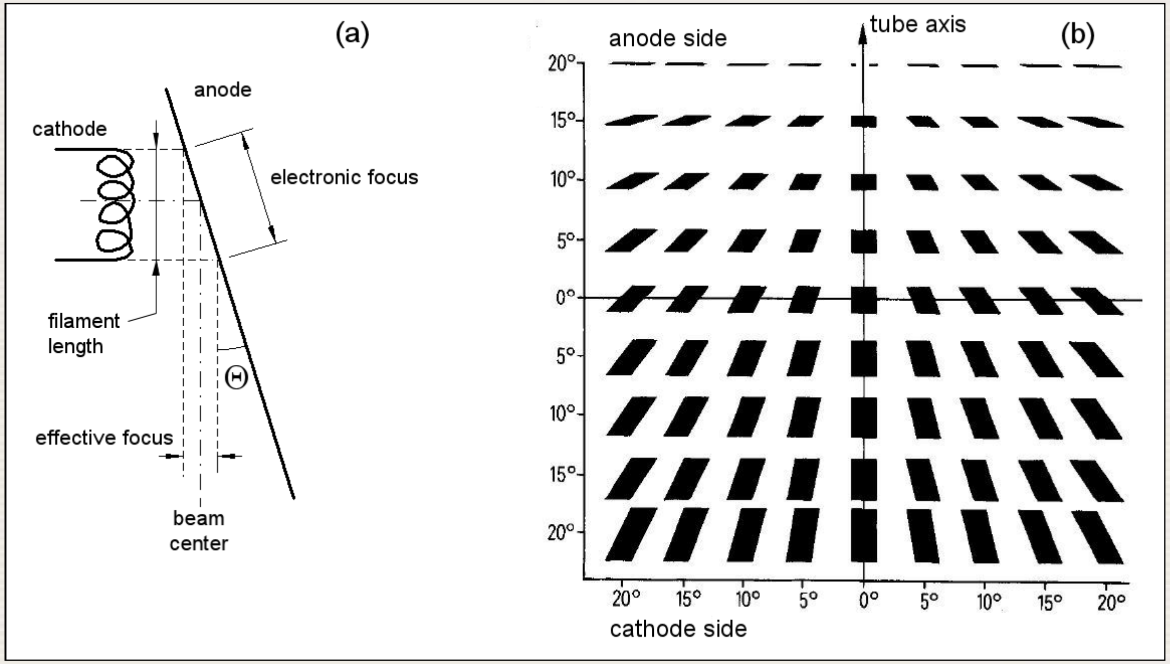
Line-Focus Principle (Anode Angle)



(a) **Line Focus Principle**: the length of the filament appears shortened in the beam direction

(b) **Graphic** representation of the focal spot shape at different locations in the radiation field (anode angle 20°)

Line-Focus Principle (Anode Angle)

The anode is **Inclined** to the tube axis typically with the **Central Ray** of the X ray field perpendicular to the tube axis

The electrons hit the anode in the electronic focus largely determined by the **Length** of the cathode filament

The electronic focus appears shortened in beam direction by **sin θ** as the **Effective Focus**

Anode angles in diagnostic tubes range from 6-22° depending on their task with 10-16° used for general purpose tubes

Line-Focus Principle (Anode Angle)

The **Radial Dimension** of the focus size is given by the filament coil diameter and the action of the focusing cup

The **Size** of the focal spot of an XRT is given for the central beam in the X ray field running **Perpendicular** to the electron beam or the tube axis

The **Actual** focal spot size depends on the position within the field of view increasing from the anode side of the tube to the cathode

Line-Focus Principle (Anode Angle)

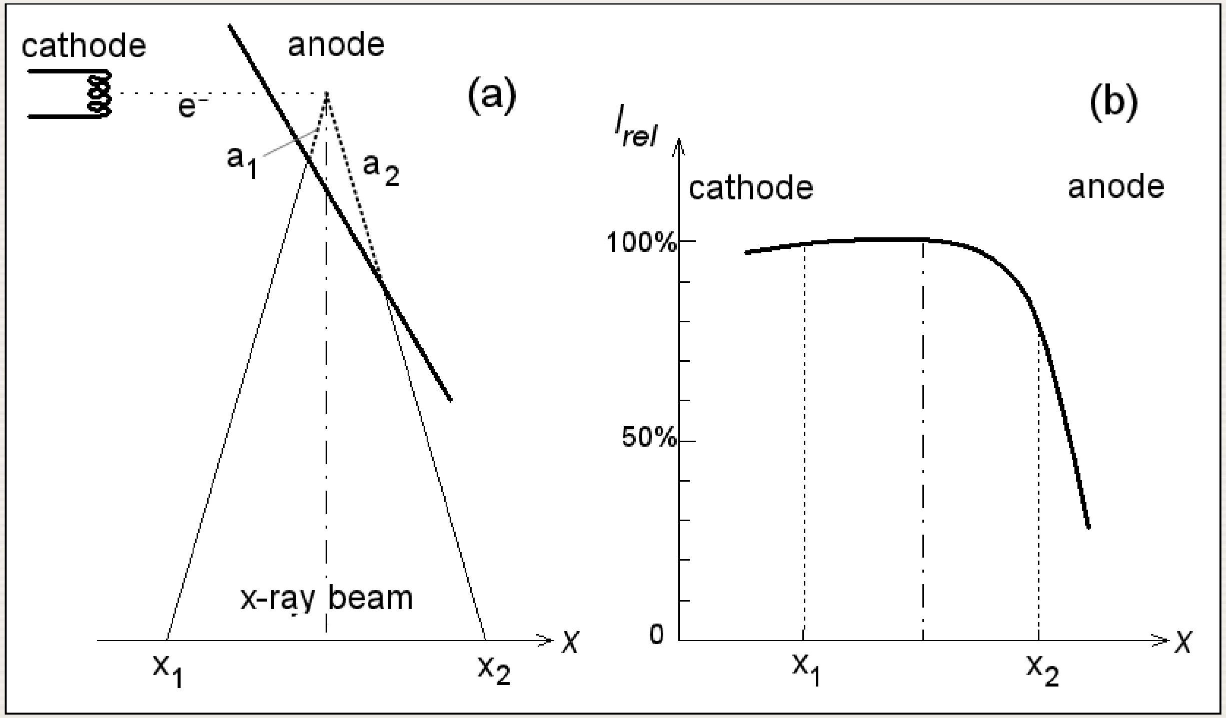
The **Reduction** of anode angles to achieve smaller effective focus sizes is limited by the size of the field of view required as the X ray beam is **Cut Off** by the anode

A further limit is given by the **Heel Effect**

Here the X rays produced at **Depth** within the anode suffer some absorption losses according to the distance passed in the anode material

Line-Focus Principle (Anode Angle)

For X rays emerging near the anode side of the X ray field the losses are higher resulting in an **Inhomogeneous** X ray intensity across the beam



(a) Absorption of X rays at the cathode side of the X ray field (**a_1**) is less than at the anode side (**a_2**)

(b) The steep drop in intensity **I_{rel}** at the anode side reflects the increased absorption (**Heel Effect**)

Line-Focus Principle (Anode Angle)

In projection radiography the heel effect can be **Verified** by measuring the air kerma across the beam

However is often **Barely Noticeable** in radiographs

In **Mammography** the heel effect is used to create a decrease in the incident air kerma from chest wall to nipple matching the decrease in organ thickness

Line-Focus Principle (Anode Angle)

In addition to X rays produced in the primary focus, some **Off-Focus Radiation** results from electrons scattered from the anode which are then accelerated back and hit the anode outside of the focal area

Extra Focal Radiation can contribute up to ~10% of the primary X ray intensity

Since the effective focal spot size for off-focus radiation is substantially **Larger** than for the primary focus it has an impact on image quality such as background fog and blurring

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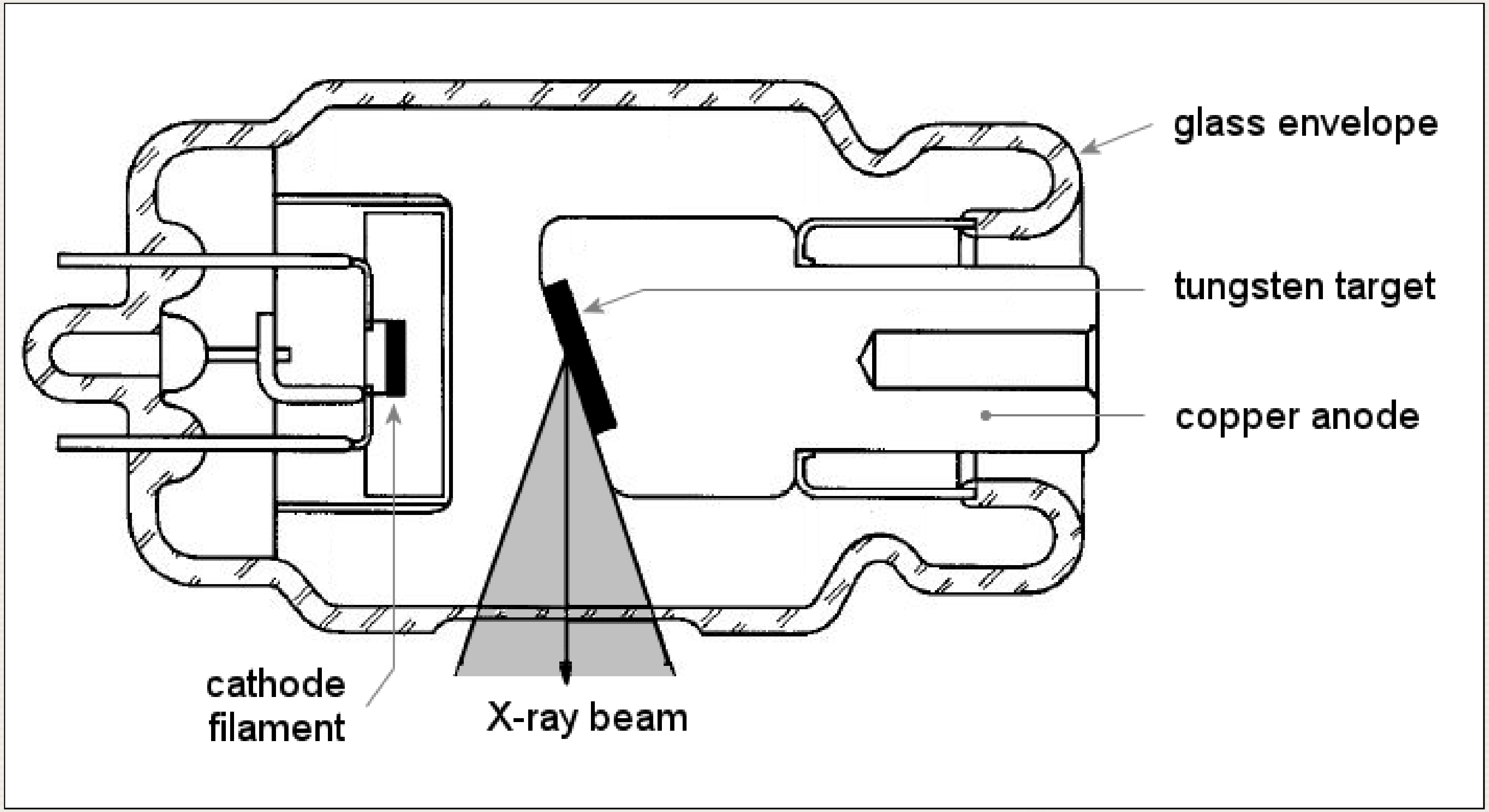
Stationary & Rotating Anodes

For X ray examinations that require only a low anode current or infrequent low power exposures (e.g. dental units, portable X ray units and portable fluoroscopy systems) an X ray tube with a **Stationary Anode** is applicable

Here a small tungsten block serving as the target is **Brazed** to a copper block to dissipate the heat efficiently to the surrounding cooling medium

As the focal spot is **Stationary** the maximum loading is determined by anode temperature and temperature gradients



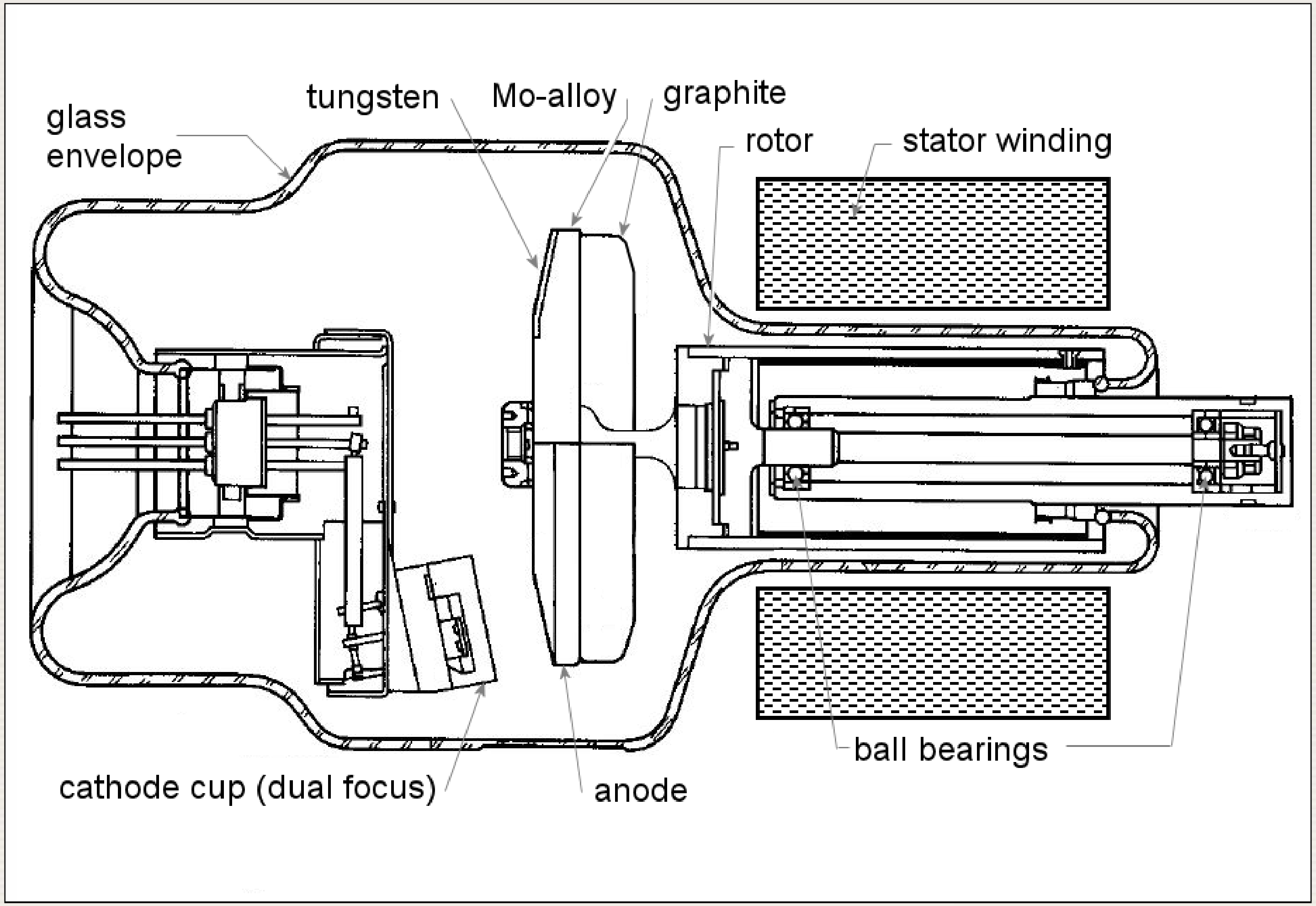
Dental XRT with a **Stationary Anode**

Stationary & Rotating Anodes

Most X ray examinations need photon fluences which **Cannot** be obtained with stationary anodes as bombarding the same spot with higher anode currents leads to **Melting** and **Destruction** of the anode

In a tube with a **Rotating Anode** a tungsten disk rotates during an exposure thus effectively increasing the area bombarded by the electrons to the circumference of a **Focal Track**

The energy is dissipated to a much larger volume as it is **Spread Over** the anode disk



XRT with a **Rotating Compound Anode**

Stationary & Rotating Anodes

The anode disk is fixed to a **Rotor** and a **Spindle** with a short **Stem**

The spindle is supported by two **Ball Bearings**

In newer developments floating bearings with **Liquid Metal** have been developed

The rotating anode is attached to the rotor of an asynchronous **Induction Motor**

The **Rotor** is mounted within the tube housing on bearings (typically ball bearings)

Stationary & Rotating Anodes

The **Squirrel-Cage** rotor is made up of bars of solid copper that span the length of the rotor

At **Both Ends** of the rotor the copper bars are connected through rings

The driving magnetic fields are produced by **Stator** windings outside the tube envelope

Stationary & Rotating Anodes

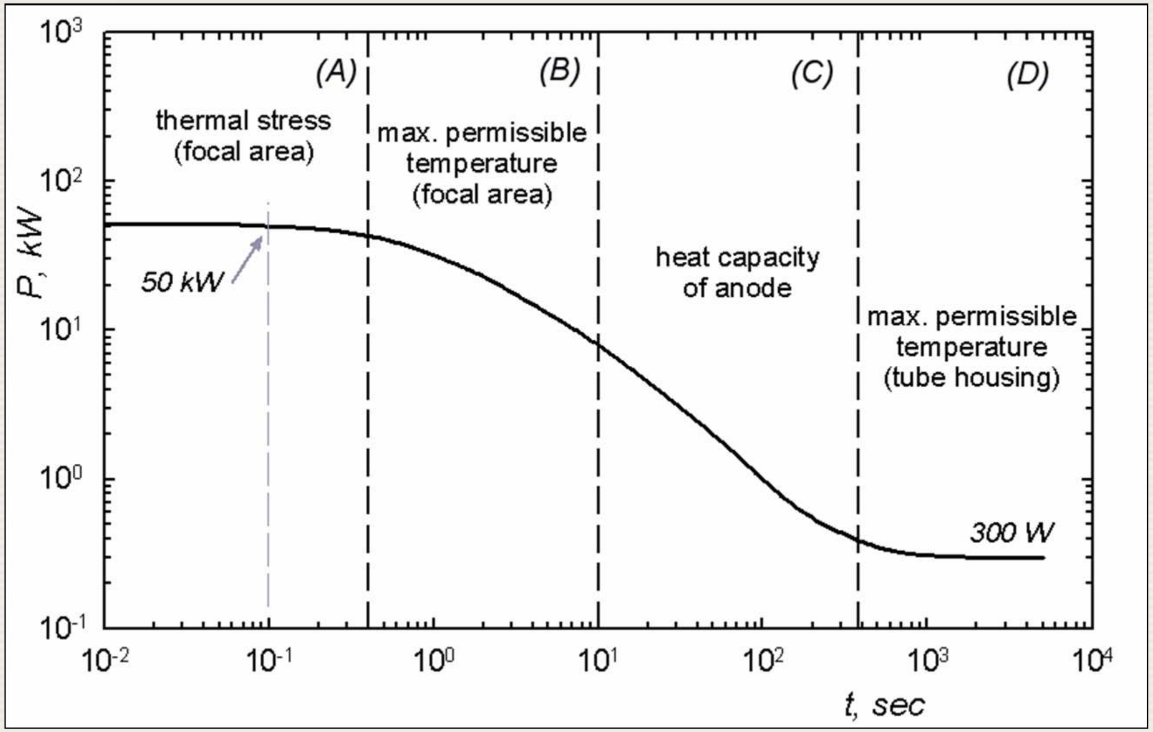
The **Rotational Speed** of the anode is determined by the frequency of the power supply and the number of active windings in the stator

Speed can be varied between high (9000-10000 rpm) and low speed (3000-3600 rpm) using all three or one phase only

Rotor Bearings are critical components of a rotating anode tube and along with the whole assembly, cycling over a large temperature range results in high thermal stresses

Thermal Properties

The **Limiting Factor** in the use of X ray tubes is given mainly by the thermal loading capacity of the anode



Maximum Permissible Tube Load vs. time for a single exposure, constant current, 100 kV tube voltage and a 50 kW tube

The **Nominal Power** is determined for an exposure time of 0.1 s

Thermal Properties

Within the **First 100 ms** the maximum load is determined by mechanical stress in the anode material developing from temperature gradients near the surface of the focal spot (**A**)

As a consequence cracks can develop leading to an increase in anode surface roughness

This effect can be reduced by:

- use of a more ductile alloy as the focal track (e.g. **Tungsten/Rhenium** alloys) or
- an increase in the size of the focal spot or the rotational speed of the anode

Thermal Properties

The energy released in the focal spot **Raises** the temperature to a max permissible level (2757°C for tungsten) for exposures up to a few seconds thus limiting the maximum load (**B**)

In CT and fluoroscopic procedures **Longer** exposure times are needed (10 s to >200 s)

- Here the dissipation of heat across the **Entire** anode disk becomes important
- The important physical properties are then the **Heat Conduction** and **Heat Capacity** of the anode disk (**C**)

Thermal Properties

The **Heat Capacity** is the energy stored in the anode disk with the anode at its maximum permissible temperature

It depends on the **Specific Heat** and **Mass** of the anode materials

Molybdenum is superior to **Tungsten** in this respect

Increasing the mass of the anode (diameter, thickness) has its limitations as **Balancing** the rotating anode becomes difficult for the wide range of temperatures occurring

Thermal Properties

Since **Graphite** has a higher specific heat at higher temperatures than molybdenum or tungsten the heat capacity can be increased by attaching graphite heat sinks to the back of the anode disk

Graphite enhances the dissipation of heat by **Black-Body Thermal Radiation**

The **Maximum Permissible Load** for long or continuous exposures is determined by the effectiveness of heat removal from the anode (**D**)

Thermal Properties

Most of the heat is removed by **Thermal Radiation** and absorbed in the tube envelope and the surrounding insulating oil

The maximum permissible temperature and the heat capacity of the **Tube Housing** is then the limiting factor of applicable power

Thermal Properties

Heat Conduction in traditional tubes from anode disk via stem, spindle, bearings and bearing support is **not** very efficient

In some tube designs the ball bearings have been replaced by special bush bearings (spiral grooves bearings) with a **Liquid Gallium Alloy** for lubrication

The **Thermal Resistance** of such bearings is much lower compared to ball bearings which enhances the heat flow and increases the continuous power rating of the tube

Thermal Properties

In the latest generation of X ray tubes (**Rotational Envelope Tube**) the removal of heat from the anode is increased considerably by directly exposing the back of the anode disk to the cooling oil

Enabling **Long** exposures with **High** anode currents as required in CT scans

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Tube Envelope

The tube envelope maintains the required **Vacuum** in the XRT

A **Failing** vacuum due to leakage or degassing of the materials causes increased ionization of the gas molecules which slows down the electrons

Further, a current of **Positive Ions** flowing back could impair or destroy the cathode filament

The envelope is commonly made of glass but high performance tubes increasingly have **Glass/Metal** or **Ceramic/Metal** envelopes

Tube Envelope

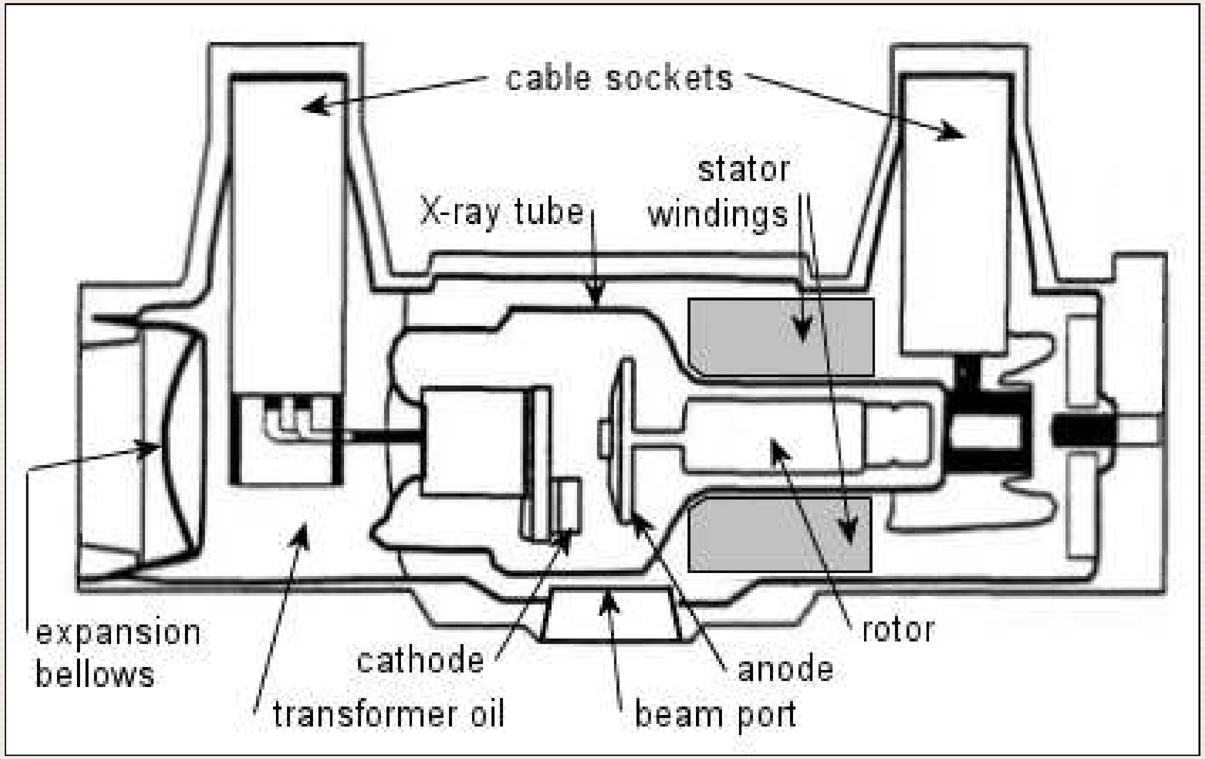
The X ray beam exits the tube through a **Window** in the envelope

To reduce absorption the **Thickness** of the glass is reduced in this area

If low-energy X rays are used as in mammography the exit port is a **Beryllium** window which has less absorption than glass due to its low atomic number

Tube Envelope

The XRT (often referred to as the **Insert**) is installed in a **Housing** providing the structural support required



Typical housing assembly for a general purpose XRT

Tube Envelope

The space between housing and envelope is filled with **Transformer Oil** serving as electrical insulation and for heat removal from the envelope surface which is heated by the infrared radiation from the anode

- The change of the oil volume with varying temperature is taken care of by the **Expansion Bellows**
- The oil carries the heat away to the housing by convection sometimes enhanced by **Forced Cooling** with a ventilator or heat exchangers

Tube Envelope

The housing also provides **Radiation Shielding** to prevent any radiation except the primary beam from leaving the housing

The inside of the housing is lined with **Lead Sheets** to minimize leakage radiation

The maximum acceptable exposure due to **Leakage Radiation** is limited by regulation

Tube housings also provide **Mechanical Protection** against the impact of envelope failure

5.4 ENERGIZING & CONTROLLING THE XRT

The **X Ray Generator**

- **Provides** all electrical power sources and signals required for the operation of the X ray tube
- **Controls** the operational conditions of X ray production
- **Controls** the operating sequence of exposure during an exam

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5.4 ENERGIZING & CONTROLLING THE XRT

The essential components are:

- a **Filament Heating** circuit to determine anode current
- a **High Voltage** supply
- a **Motor Drive** circuit for the stator windings required for a rotating anode tube
- an **Exposure Control** providing the image receptor dose required
- an **Operational Control**

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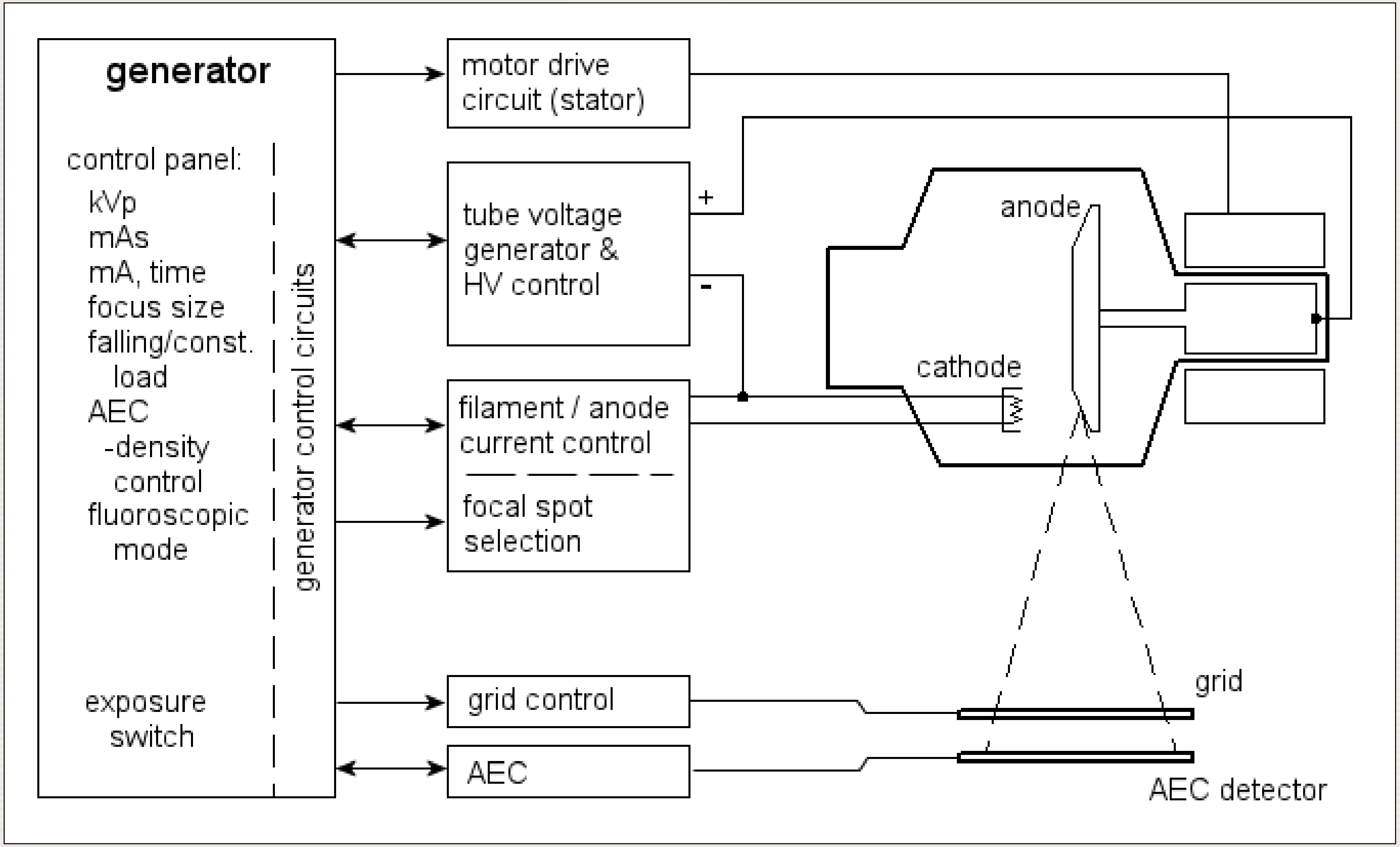
5.4 ENERGIZING & CONTROLLING THE XRT



Schematic diagram of a basic X ray generator

5.4 ENERGIZING & CONTROLLING THE XRT

The **Operational Control** is often accomplished by a microprocessor system but electromechanical devices are still in use

Modern generators provide control of the **Anode Temperature** by

- **Monitoring** the power applied to the tube and
- **Calculating** the cooling times required according to the tube rating charts

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5.4 ENERGIZING & CONTROLLING THE XRT

5.4.1 Filament Circuit

An **Isolated Transformer** supplies the filament heating current

The generator is **Programmed** to set the heating current according to the tube characteristics

Heating currents range up to 10 A with voltages of <15 VAC

To minimize thermal stress and increase durability, the filament is **Permanently Preheated** to a temperature for which thermionic emission is negligible

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5.4 ENERGIZING & CONTROLLING THE XRT

5.4.1 Filament Circuit

The **Thermal Inertia** of the filament limits the speed of change in tube current (e.g. falling load)

Thermal Time Constants range from 50-200 ms

For a frequency of heating currents of 100 or 120 Hz some tube current **Ripple** is due to the temperature variations induced

For high frequency generators the thermal inertia of the filament suppresses **Fluctuations** of thermionic emission

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Irrespective of the waveform the tube voltage is defined as the **Peak Voltage**, kVp, of the voltage train

The **Voltage Ripple**, R, is given as the relative difference of the minimum voltage, kV_{min}, from the peak voltage:

$$R = (kVp - kV_{min}) / kVp$$

5.4 ENERGIZING & CONTROLLING THE XRT
5.4.2 Generating the Tube Voltage

The tube voltage is supplied **Symmetrically** to the tube, i.e. a net potential difference of 150 kV is achieved by feeding -75 kV to the cathode and +75 kV to the anode

This is electrically accomplished by **Grounding** the centre tap of the secondary coil of the high voltage transformer

Requirements for electrical isolation are less stringent then

In **Mammography** with tube voltages <40 kV and with some high performance tubes one electrode is kept at ground potential

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Except for grid controlled tubes the length of an exposure is determined by the provision of high voltage to the tube by switching in the **Primary** circuit

Electromechanical Relays were employed in single- and three-phase generators, but now electronic switching components, such as **Thyristors**, are used

Timing in single-phase generators is only possible in **Multiples** of pulses giving inaccurate timing for short exposures

Three-Phase Generators use a pre-pulse of low current to avoid magnetic saturation of the transformer core

When the high voltage is turned off the **Charge Stored** in the cable capacitance and the circuit is discharged via the XRT

The end of the voltage waveform therefore shows some **Tailing**, an effect impairing the production of short pulses

Single-Phase Generators

Single-phase generators use a single phase mains supply and a **Step Up Transformer** with a fixed winding ratio

The high voltage is set by a variation of the primary voltage with a switched **Autotransformer**

Half-Wave Rectification of the transformed voltage gives a **1-Pulse** waveform where a pulse is a half-wave per period of mains frequency (50 or 60 Hz)

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Single-Phase Generators

Some low-power X ray units use the tube as a **Self Rectifying Diode** with current only flowing from the cathode to the anode

but reverse current flow, as a result of a **Hot Anode** is a limiting factor

Today **Solid-State Diodes** are used as rectifiers

A **Full-Wave Rectification** yields two half-waves per period (**2-Pulse** waveform)

Voltage Ripple of 1- and 2-pulse waveforms is **100%**

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Three-Phase Generators

With a three-phase mains supply three AC-voltages each with a **Phase-Shift** of 120° are available

Full-Wave Rectification gives then 6 half-waves per period (**6-Pulse** waveform) with a nominal ripple of 13.4%

Due to imbalances in transformer windings and voltages the ripple might in practice approach **25%**

Adding another **Secondary Winding** to the transformer gives two secondary voltages

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Three-Phase Generators

Combining the full-wave-rectified secondary voltages using **Delta**- and **Wye-Connections** yields a total of 6 phases with a phase shift of 60° each

Full-wave rectification then gives a total of **12 Pulses per Period** with a nominal ripple of 3.4% (in practice <10% is achieved)

Three-phase generators are **More Efficient** and allow for much higher tube output than single phase

High-Frequency Generators

This type of generator includes a **Stabilized Power Supply** in the front end of the device

First the **Mains** supply is rectified and filtered to produce a DC-supply voltage needed for an **Inverter Circuit**

The **Inverter** generates pulses which are transformed, rectified and collected in a capacitor to give the high voltage for the tube

The inverter **Pulse Rate** is used to control the tube voltage

High-Frequency Generators

The actual voltage on the tube is sensed by the generator and compared with the voltage set on the console

The difference then is used to change the **Pulse Rate** of the inverter until the set voltage is achieved

Similarly a separate inverter system is used for the tube current

The **Pulse Shape** of a single X ray exposure pulse resembles a fundamental frequency of several tens of kHz giving rise to the generator's name

High-Frequency Generators

Transformers for such frequencies are much **Smaller** than for 50/60Hz voltages reducing the size and weight substantially

In low-power generators the whole generator could be included in the tube housing avoiding any high-voltage cabling

The **Voltage Ripple** depends on many technical factors but for low-power applications is typically **~13%**, dropping to **~4%** at higher currents

The **Time Constants** relevant for voltage and current control are typically **<250 µs** enabling better timing control of the exposure than with single and three-phase generators

Capacitive Discharge Generators

In places with inadequate mains supply or in remote locations capacitor discharge generators are helpful

A capacitor is **Charged** to a high voltage just before an exposure

The capacitor is connected to the XRT with the start and length of exposure controlled by a **Grid**

High tube currents and short exposure times can be obtained

However, discharging a capacitor implies a **Falling Tube Voltage** during exposure

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Capacitive Discharge Generators

Typically **Voltage Drops** of ~1 kV per mAs are usual

As kerma drops with voltage the appropriate exposure of thick body parts can be problematic

Constant Voltage Generators

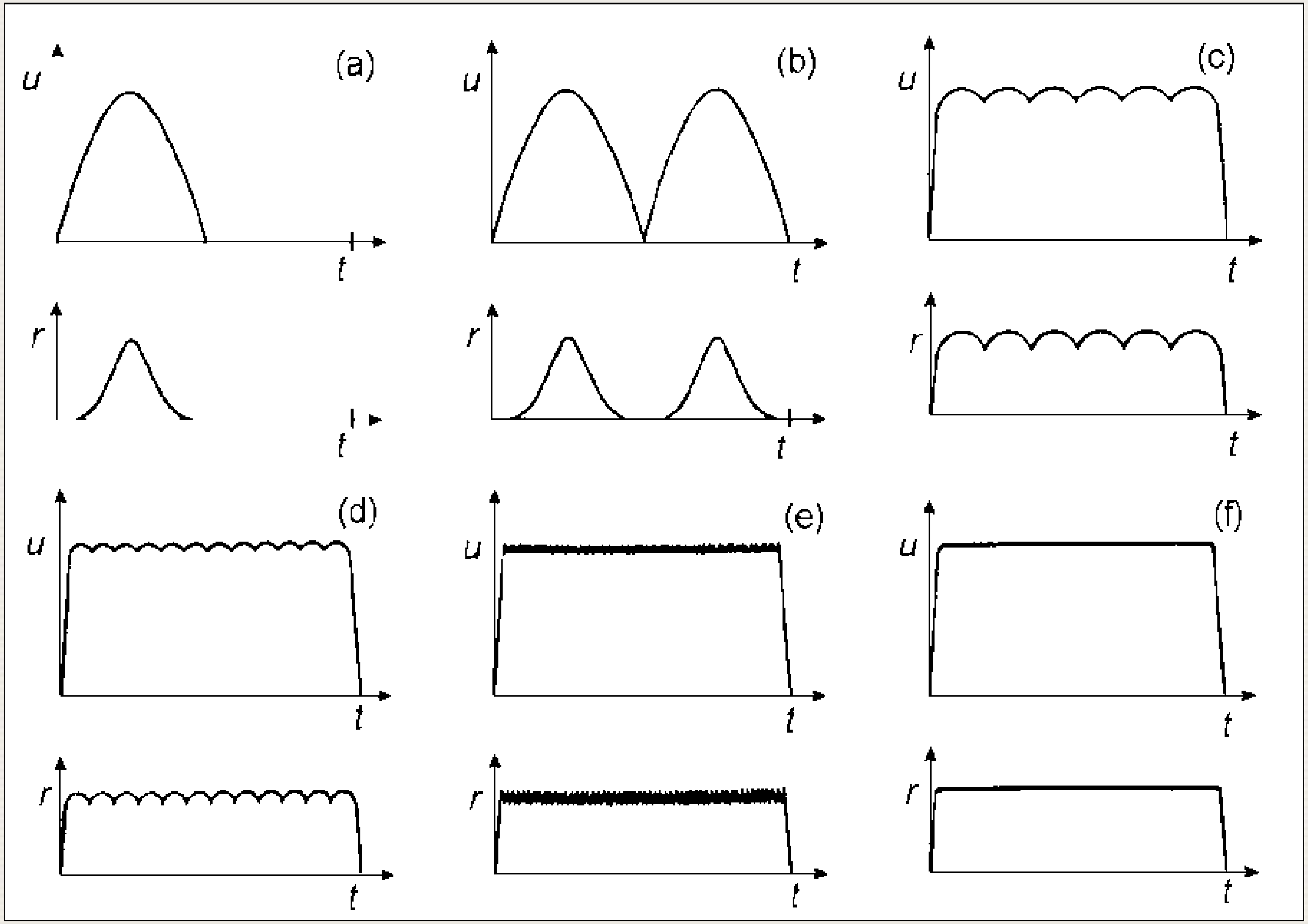
Constant voltage generators achieve a DC-high voltage with minimal ripple through the use of a **Closed Loop Linear Voltage Controller** (e.g. high-voltage triodes) in series with the tube

High frame rates and voltage stability are achieved

Constant potential generators use a **Complex** technology with high costs of investment and operation, and consequently have lost popularity

5.4 ENERGIZING & CONTROLLING THE XRT

5.4.2 Generating the Tube Voltage



Voltage waveforms **u** and associated tube output (dose rate) **r** for

(a) 1-pulse

(b) 2-pulse

(c) 6-pulse

(d) 12-pulse

(e) high-frequency

(f) constant voltage generators

Comparison of Generator Technologies

In radiology it is desirable to keep exposure times

As Low As Achievable

1-pulse waveforms produce radiation in only half of a cycle, **Double** the exposure time compared to 2-pulse voltages

As the kerma output rises approximately with the **Square** of the tube voltage there is a substantial amount of time in a half-wave of 1- and 2-pulse waveforms with little or no contribution to kerma output, again effectively increasing the exposure time

Comparison of Generator Technologies

The 1- and 2-pulse waveforms also yield **Softer** X ray spectra which implies increased radiation dose to the patient

Both exposure time and patient dose indicate that the optimum waveform would be a DC-voltage with essentially no ripple but 12-pulse and high-frequency generators are **Near Optimum**

Comparison of Generator Technologies

Generators transforming mains AC-voltages suffer from external voltage instabilities

Devices for compensating for these fluctuations are often integrated into the generator design, however **High Frequency Generators** that provide tube supplies with higher stability and accuracy, are currently the state-of-the-art

5.4 ENERGIZING & CONTROLLING THE XRT

5.4.3 Exposure Timing (AEC)

Exposure of a radiograph can be set **Manually** by choosing tube current and exposure time

Except in examinations with little variability in body dimensions (e.g. extremities) an **Automatic Exposure Control** (AEC) is mandatory to achieve a consistent image quality or film density

The AEC **Terminates** an exposure when the image receptor has received a **Preset** level of radiation

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5.4 ENERGIZING & CONTROLLING THE XRT

5.4.3 Exposure Timing (AEC)

The AEC-system consists of **1-3** radiation detectors (ionization chambers or solid-state detectors)

The signal of these detectors is **Amplified** and integrated, corrected for response in photon energy and dose rate, and finally **Compared** to the preset dose level

The exposure is terminated when the **Chosen Level** is attained

In case the AEC does not terminate the exposure a **Backup Timer** sets a time limit

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On **Installation** of a radiographic unit the dose levels are set taking into consideration all the components of the imaging chain, i.e. film and screens, imaging plates, film development, preferred tube voltage and filtration, acceptable image noise, etc

This process needs to be carried out for **All** tube voltages, image receptor and examination types in question

Some products allow for fine manual adjustment to the pre-set dose level by a **Density Control** on the console adapting the density in steps of 10-20%

5.4.3 Exposure Timing (AEC)

Radiographic devices commonly have **Ionization Chambers** as AEC detectors positioned immediately in front of the radiographic cassette

The detectors must show no visible radiographic contrast on the image

For **Low-Energy** X ray units (e.g. mammography, paediatric units) this is difficult to achieve and detectors are positioned behind the image receptor

Solid-State Detectors are mostly employed in this case

5.4 ENERGIZING & CONTROLLING THE XRT

5.4.3 Exposure Timing (AEC)

The position of the detectors is **Delineated** on the table top or wall stand to assist the operator in patient positioning

As absorption in the patient's body can vary substantially across the beam the operator can **Select** a detector or a combination of detectors for exposure control to obtain optimal exposure in the dominant part of the image

As an example, for a chest X ray in PA projection, the **Two Lateral** detectors, positioned under the lung regions, are chosen, while in lateral projection the **Central** detector is selected

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To avoid image blurring due to patient motion short exposure times are **Mandatory**

To produce the shortest possible exposure the generator starts with the maximum permissible current and in the course of the exposure lowers the tube current consistent with tube ratings (**Falling Load**)

Thus the tube is operating at the **Maximum** permissible power rating during the entire exposure

5.4 ENERGIZING & CONTROLLING THE XRT
5.4.4 Falling Load

In some products an exposure with falling load can be run at a reduced power setting (e.g. 80 % of the maximum power) to lower the stresses

The operator sets tube voltage, focus size and if not in AEC-mode the **mAs**-value, but not mA and time

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5.5 XRT & GENERATOR RATINGS

5.5.1 X Ray Tube

The **Nominal Voltage** gives the **Maximum** permissible tube voltage

For most tubes this will be **150 kV** for radiography

For fluoroscopy another nominal voltage might be specified

The **Nominal Focus** is a dimensionless figure characterizing the focal size (IEC336)

For each nominal focus a range of **Tolerated** dimensions is given for width and length of the focus

e.g. a **Nominal Focus** of 1.0 allows for a width of 1.0-1.4 mm, and a length of 1.4-2.0 mm

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5.5 XRT & GENERATOR RATINGS

5.5.1 X Ray Tube

The **Power Rating** P for a given focus is the maximum permissible tube current for a 0.1 s exposure at a tube voltage of 100 kV

A more practical quantity is the power rating obtained with a thermal preload of the anode (**Thermal Anode Reference Power**) of typically 300 W

P depends on focal spot size and ranges from **~100 kW** for 1.5 mm down to **~1 kW** for 0.1 mm focus size

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5.5 XRT & GENERATOR RATINGS

5.5.1 X Ray Tube

Physical data for the anode includes target **Angle**, anode **Material** and **Diameter** of the disk

The anode drive frequencies determine the **Rotational Speed** of the anode

High Power Loading of the anode requires high rotational speeds

To prevent the bearings from wear and damage, the speed is reduced for low power settings as in **Fluoroscopy**

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The **Heat Capacity**, Q, of the anode is the heat stored in the anode after arriving at the maximum permissible temperature

Q is equivalent to an electrical energy of

$$Q = \mathbf{w} \cdot U_A \cdot I_A \cdot t$$

with tube voltage U_A and current I_A, and t the exposure time

U_A given as a peak voltage is multiplied with a waveform factor **w** to obtain the effective tube voltage (RMS voltage)

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5.5 XRT & GENERATOR RATINGS

5.5.1 X Ray Tube

Waveform Factor, **w** equals:

- 0.71 for 1- and 2-pulse generators
- 0.96 for 6-pulse
- 0.99 for 12-pulse generators

Q is then given in **Joule** (J)

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5.5 XRT & GENERATOR RATINGS

5.5.1 X Ray Tube

Since early generators were based on single phase supplies **w** was simply set to **1.0** for 1- and 2-pulse and **1.35** for 6-pulse generators giving the heat capacity in another unit, the **Heat Unit** (HU), where:

$$1 \text{ J} = 1.4 \text{ HU}$$

The heat capacity of general purpose tubes starts at **~200 kJ** ranging up to **>1000 kJ** for high performance tubes

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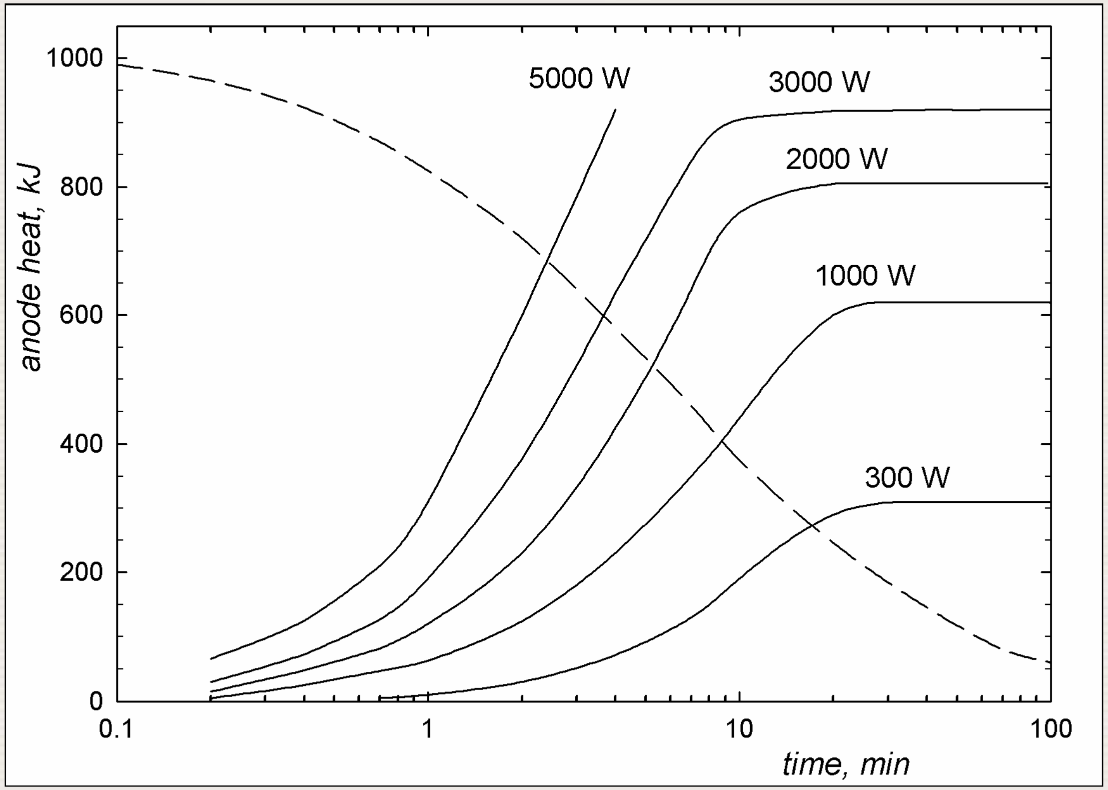
The **Maximum Anode Heat Dissipation** indicates the maximum rate of heat loss typically available at maximum anode temperature

This data depends on temperature and tube type

Tube data include also **Cooling** and heating characteristics

Cooling of the anode (dashed curve) and heat build-up for several constant input heat rates



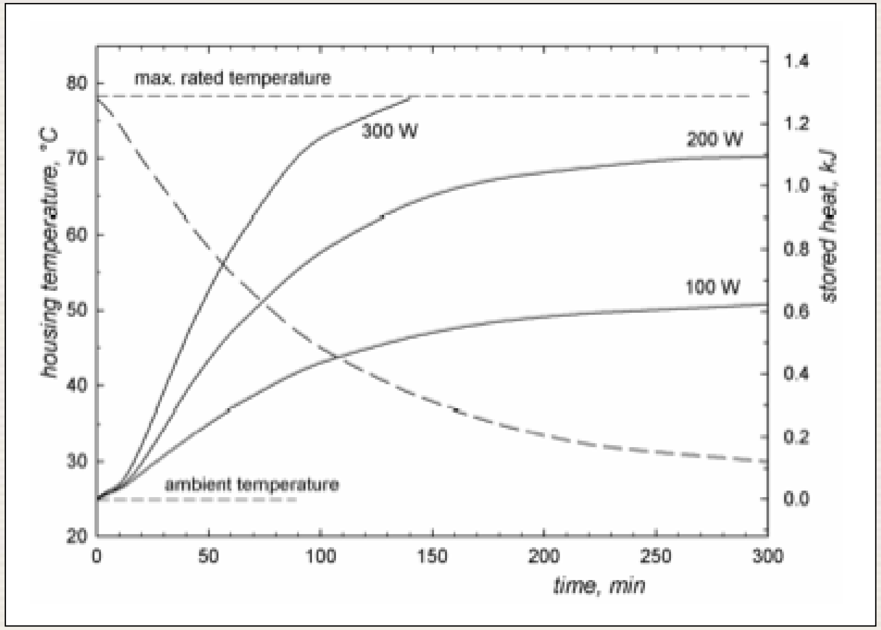
Rated heat capacity of the anode is 1000 kJ

5.5 XRT & GENERATOR RATINGS
5.5.2 Tube Housing

The **Maximum Heat Capacity** for a tube assembly is typically in the range of 1000-2000 kJ

Maximum Continuous Heat Dissipation describes the steady state of heat flowing in and cooling off

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



5.5.2 Tube Housing

The **Patterns** of loading the tube in an examination vary from single radiographic exposures to long high-current CT-scans, from simple fluoroscopic examinations to long interventional procedures

The **Tube Rating Charts** contain basic data to estimate required cooling times

These limits **Have** to be observed in particular if the control panel gives no indication on actual tube loading or required cooling times

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5.5 XRT & GENERATOR RATINGS

5.5.2 Tube Housing

Exposures made by **Physicists** in their measurements can be repeated much more frequently than within the course of a patient examination

Several such high power exposures without observation of the appropriate cooling times can **Damage** the anode and bearings

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5.6.1 Collimator & Light Field

The limitation of the X ray field to the size required for an examination is accomplished with **Collimators**

The benefits of collimating the beam are **Twofold**:

- **Reduction** of patient dose
- **Improvement** of image contrast due to reduced scattered radiation

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5.6 COLLIMATION & FILTRATION

5.6.1 Collimator & Light Field

A **Collimator Assembly** is typically attached to the tube port defining the field size with adjustable parallel-opposed lead **Diaphragms** or blades

To improve the effectiveness of collimation **Another Set** of blades might be installed at some distance to the first blades in the collimator housing

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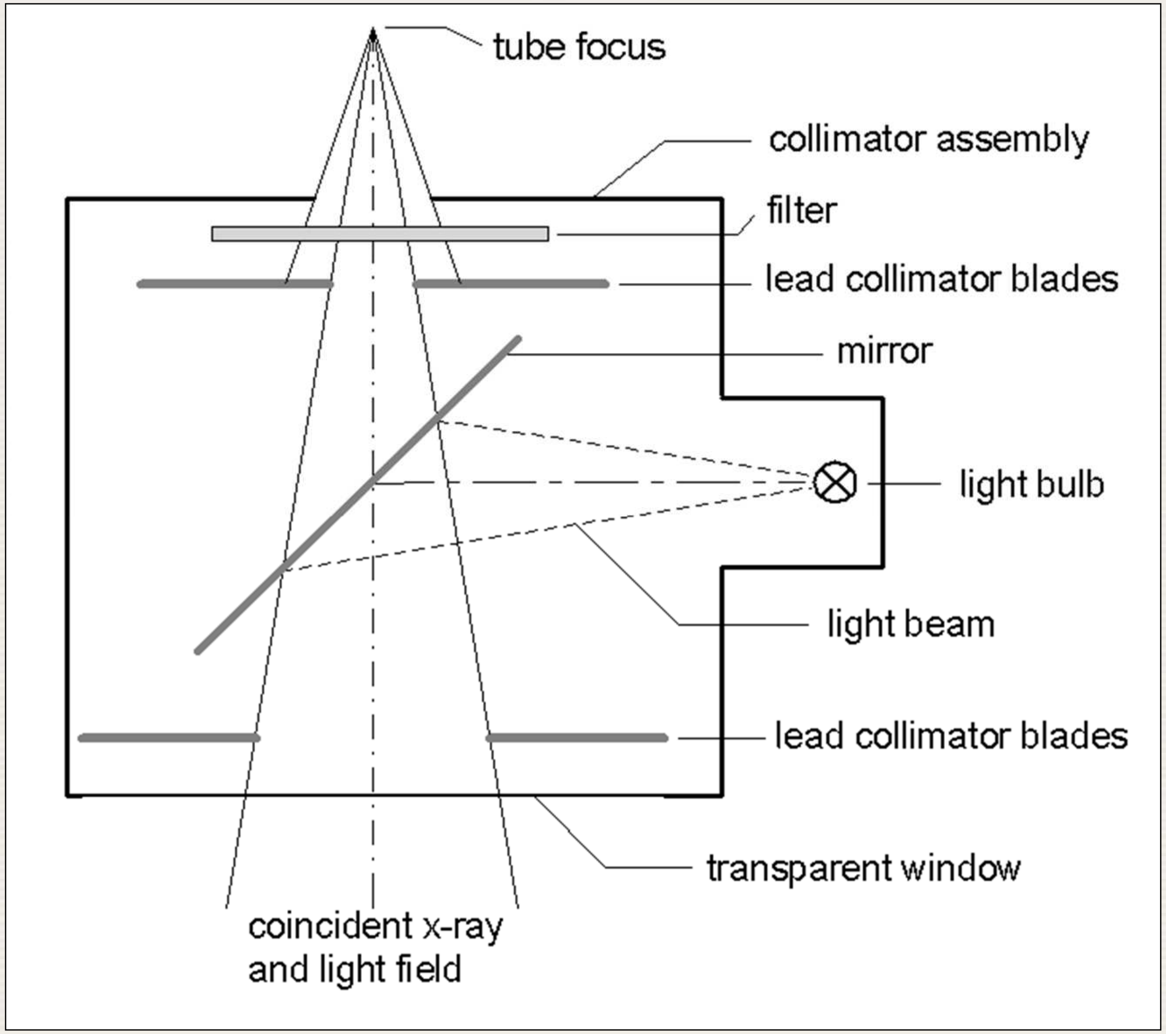
5.6 COLLIMATION & FILTRATION

5.6.1 Collimator & Light Field



Typical X ray field collimator assembly

5.6 COLLIMATION & FILTRATION

5.6.1 Collimator & Light Field

Visualization of the X ray field is achieved by a **Mirror** reflecting the light from a bulb

The bulb position is adjusted so the mirrored light appears to have the same origin as the **Focal Spot** of the tube

The light field then mimics the actual X ray field

The **Congruency** of light and X ray field is subject to quality control

One must be aware that some of the penumbra at the edges of the radiation field is due to **Extra Focal Radiation**

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Adjustment of the field size is done **Manually** by the operator

But with a **Positive Beam Limitation System** the size of the imaging detector is automatically registered and the field size is adjusted accordingly

For **Fluoroscopy** other collimator types are in use with variable circular and slit diaphragms

In some applications (dental and head examinations) **Beam Restrictors** with a fixed field size are typically used

5.6 COLLIMATION & FILTRATION

5.6.2 Inherent Filtration

X rays generated in the anode pass various **Attenuating** materials before leaving the tube housing, including:

- Anode
- Tube envelope exit port (glass or metal)
- Insulating oil
- Window of the tube housing

This **Inherent Filtration** is measured in aluminium equivalents in units of mm Al

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5.6 COLLIMATION & FILTRATION

5.6.2 Inherent Filtration

Aluminium does not perfectly mimic the atomic composition of the attenuating materials present

Thus measurement of the **Al Equivalent** is usually made at 80 kVp (or otherwise the kVp settings should be stated)

Typically the inherent filtration ranges from **0.5-1 mm Al**

The mirror and the window in the collimator housing also contribute to inherent filtration with an Al-equivalent of **~1 mm**

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5.6.3 Added Filtration

Since filtration effectively reduces the low-energy component in the X ray spectrum, a minimum **Total Filtration** of at least **2.5 mm Al** is required to reduce unnecessary patient dose

Additional Filter material is positioned between tube window and collimation assembly as required

Typical filter materials include **Aluminium** or **Copper**, and in some cases rare earth filters such as **Erbium** that utilise K-edge attenuation effects

5.6.3 Added Filtration

Individual filters may be **Manually** selected on some units

In modern fluoroscopy units filters are inserted **Automatically** depending on the examination program chosen

The **Effect** of added filtration on the X ray output is an increase in the mean photon energy and HVL of the beam

As the X rays become more penetrating less incident dose at the patient entrance is required to obtain the same dose at the image receptor giving a **Patient Dose Reduction**

5.6.3 Added Filtration

Since image contrast is higher for low-energy X rays, the addition of filters **Reduces Image Contrast** and optimum conditions must be established depending on the type of examination

Added filtration also **Increases Tube Loading** as the tube output is reduced and must be compensated for by an increase in mAs to obtain the image receptor dose required

In **Mammography** special provisions concerning filtration are required to obtain the optimum radiation qualities

5.6.4 Compensation Filters

In some examinations the range of X ray intensities incident upon the image receptor **Exceeds** the range of the detector

Compensation or **Equalization Filters** can be used to reduce the high intensities due to thinner body parts or regions of low attenuation

Such filters are usually inserted in the collimator assembly or close to the tube port

Examples of compensation filters include:

- **Wedge** filters for lateral projections of the cervical spine
- **Bow tie** filters in CT

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5.7.1 Quantities Describing X Ray Output

Total Photon Fluence is not a satisfactory quantity to describe X ray output

Rather it is the **Spectral Distribution** of the photon fluence as a function of photon energy which is useful

Spectral data are rarely available for individual X ray units although computer programs exist which give useful **Simulations**

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5.7.1 Quantities Describing X Ray Output

X ray **Tube Output** can be expressed in terms of the air kerma and measured free-in-air

A measure for the **Penetrability** and the **Quality** of the X ray spectrum is the Half-Value Layer, (**HVL**)

The HVL is the thickness of absorber needed to attenuate the X ray beam incident air kerma by a factor of **Two**

In diagnostic radiology **Aluminium** is commonly chosen as absorber giving the HVL in units **mm Al**

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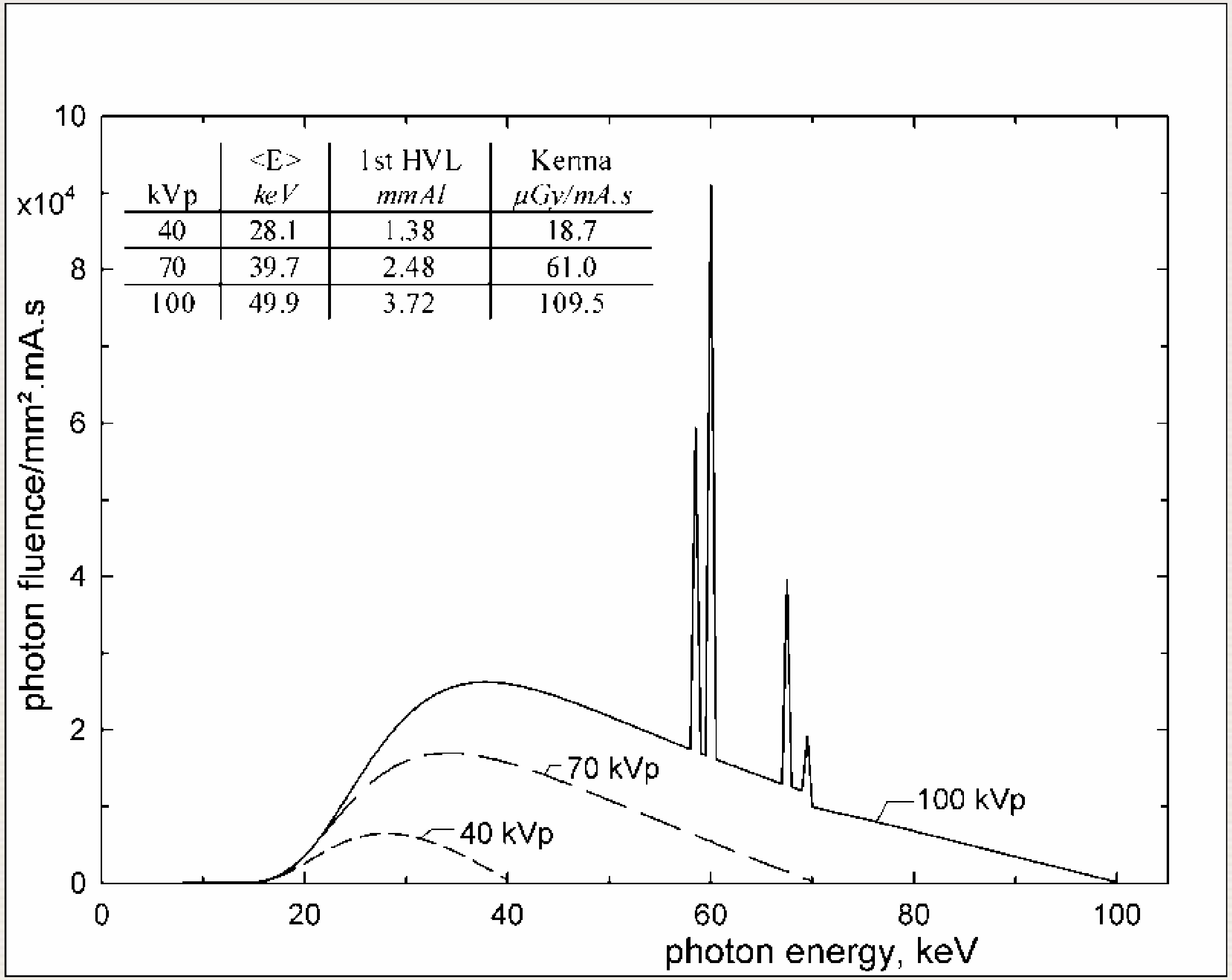
5.7 FACTORS INFLUENCING X RAY SPECTRA & OUTPUT

5.7.2 Tube Voltage & Current



X ray spectra for various tube voltages and a tungsten target
(constant voltage; anode angle 12°)

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5.7 FACTORS INFLUENCING X RAY SPECTRA & OUTPUT

5.7.2 Tube Voltage & Current

Both **Maximum** and **Mean** photon energy depend on kV

The shape of the **Low Energy** end of the spectrum is determined by the anode angle and the total filtration

Note the appearance of **Characteristic Radiation** in the 100 kV beam and the increase in **Photon Yield** with increasing tube voltage

Tube current has no influence on the **Photon Distribution**; however photon intensities are proportional to mAs

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5.7 FACTORS INFLUENCING X RAY SPECTRA & OUTPUT

5.7.3 Tube Voltage Ripple

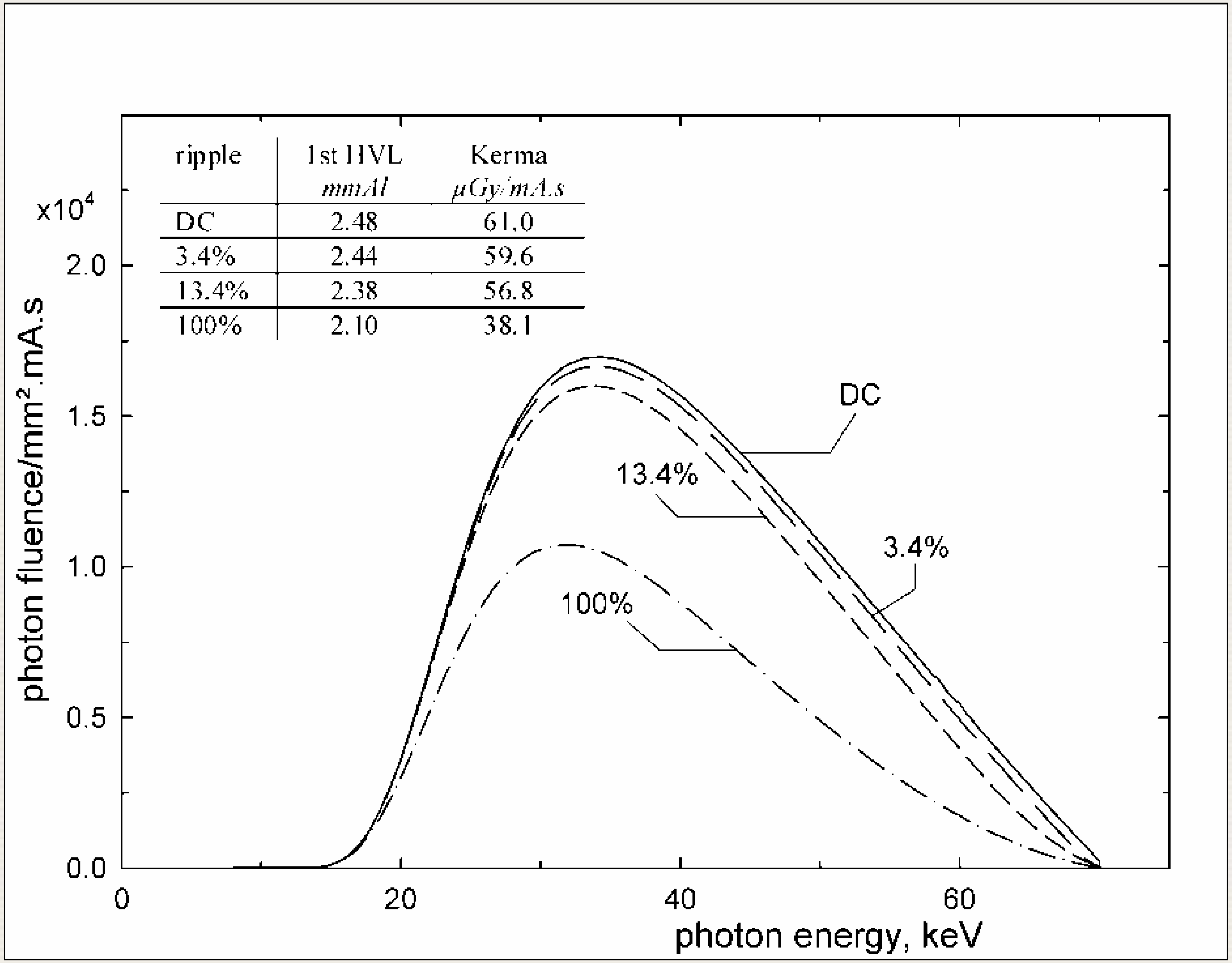
Variation of X ray spectra from a tungsten target with tube voltage ripple at 70 kVp tube voltage

DC: constant potential

3.4%: 12-pulse or converter generator

13.4%: 6-pulse generator

100%: 2-pulse generator



5.7 FACTORS INFLUENCING X RAY SPECTRA & OUTPUT

5.7.3 Tube Voltage Ripple

A DC voltage gives the **Hardest** spectrum with **Maximum** photon yield

With an increase in ripple the yield **Drops** and the spectrum **Softens**

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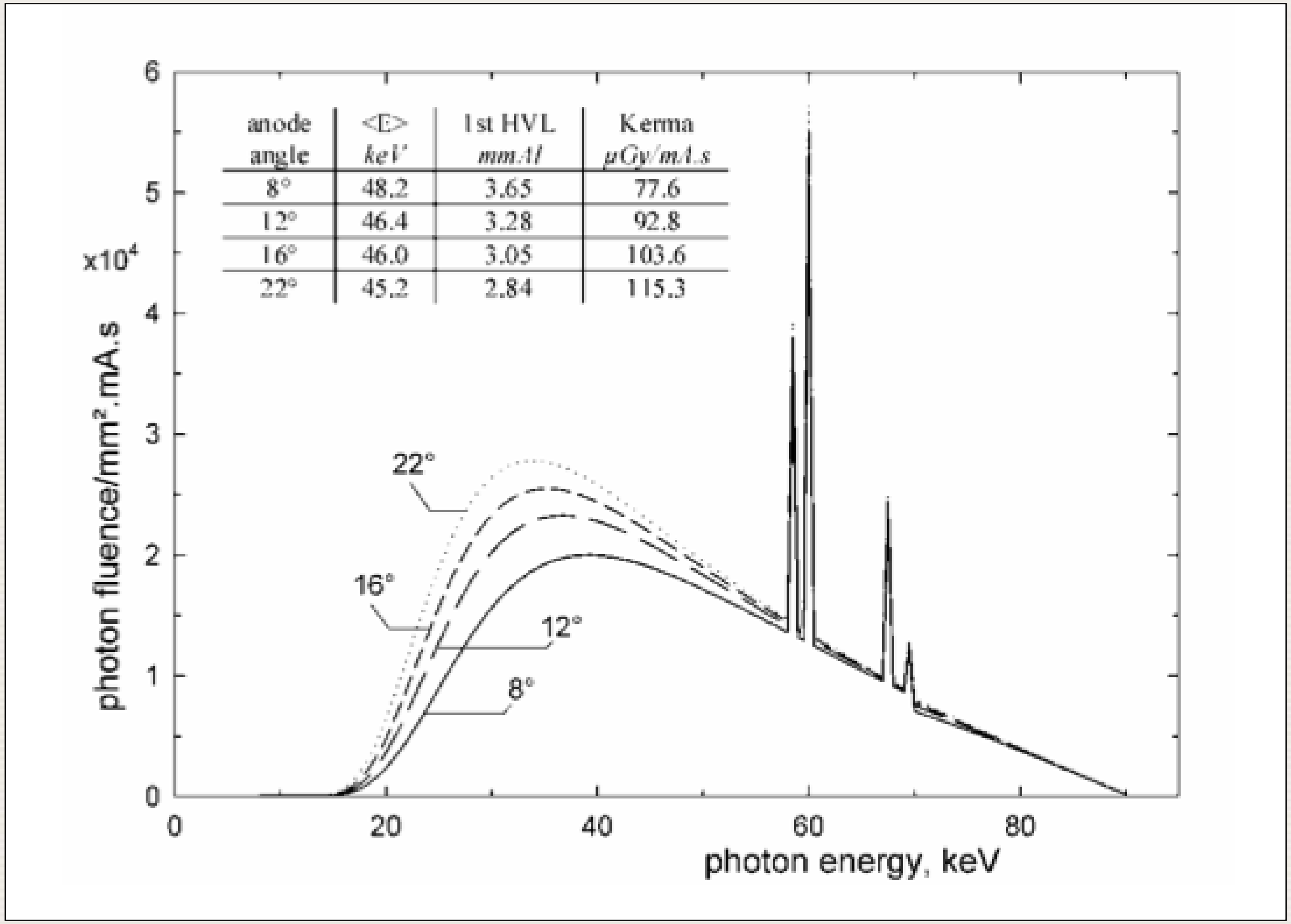
5.7.4 Anode Angle



X ray spectra obtained for various anode angles and a tube voltage of 90 kV (DC)

5.7 FACTORS INFLUENCING X RAY SPECTRA & OUTPUT

5.7.4 Anode Angle

The **Anode Angle** determines the degree of X ray absorption in the anode material

A decrease in anode angle causes an increase in the **Absorption Length** within the target

Accordingly, the maximum photon energy remains unchanged but **Hardness** increases and **Yield** drops with decreasing anode angle

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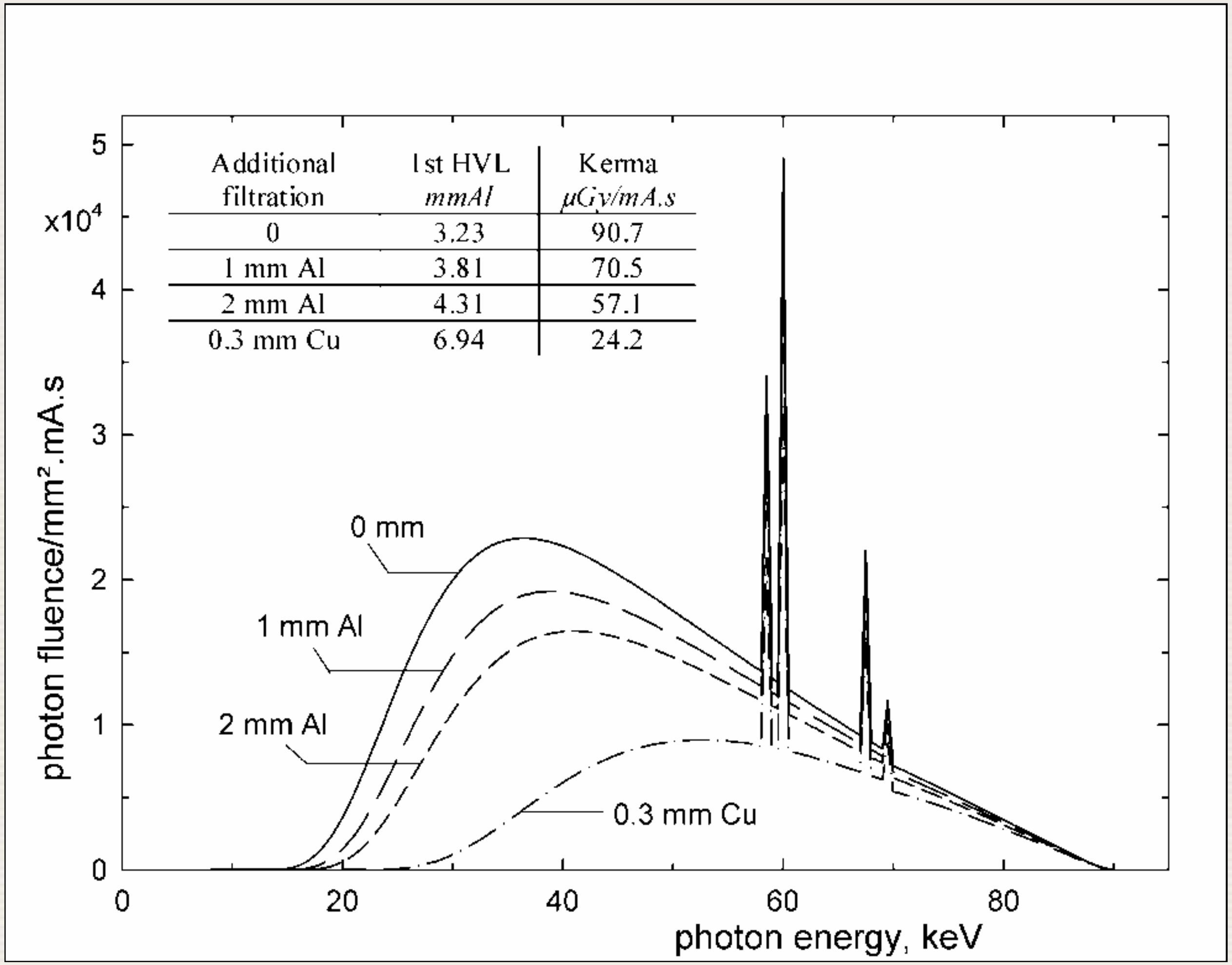
5.8 FILTRATION



Effect of additional filtration on the X ray spectrum
(90 kV, 3.4% ripple)

5.8 FILTRATION

Increasing filtration gives **Spectral Hardening** and reduction in **Tube Output**

X Ray Contrast declines with spectrum hardness which should be considered in the selection of optimal exposure parameters

Anode Roughness increases with total tube workload and increases self-filtration

Hence tubes tend to show a slight **Increase** in X ray hardness and a **Decrease** in kerma output with operational tube life

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