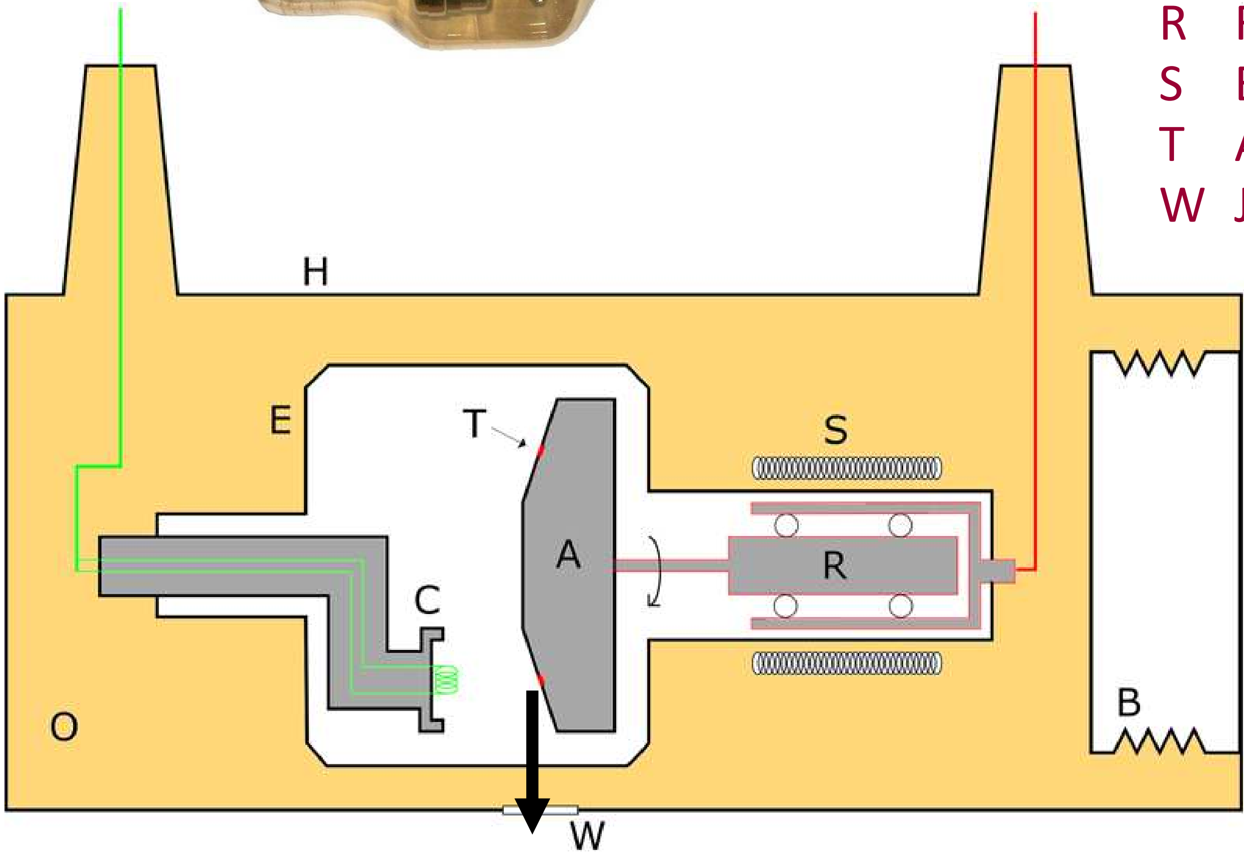
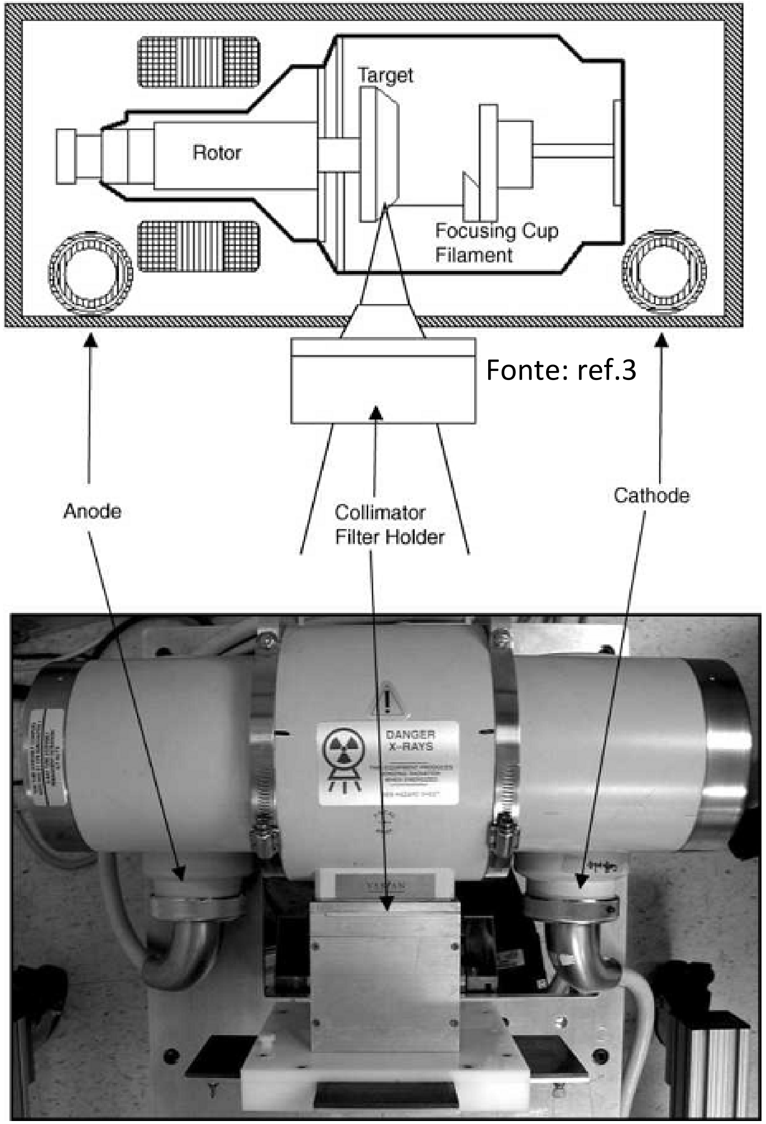
O tubo de raios-X: constituição

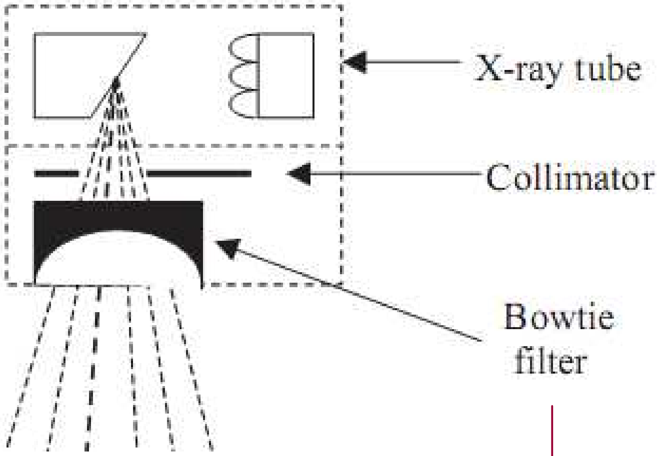


A Ânodo
B Foles de expansão do óleo
C Cátodo e filamento
E Invólucro do tubo (evacuado)
H Caixa do Tubo
O Óleo isolador
R Rotor (motor de indução)
S Estator (motor de indução)
T Alvo de tungsténio
W Janela do tubo (Al ou Be)

O tubo de raios-X: constituição



Fonte: ref.3

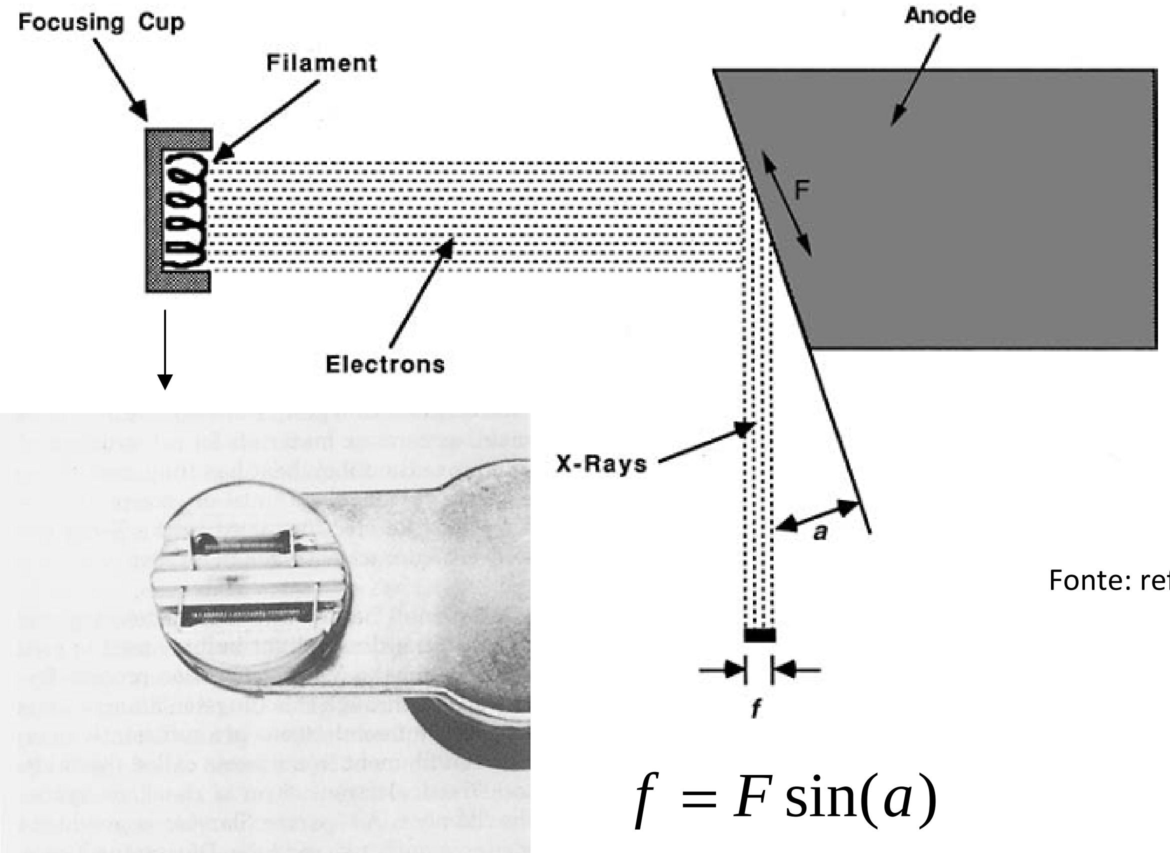


Define os limites do feixe

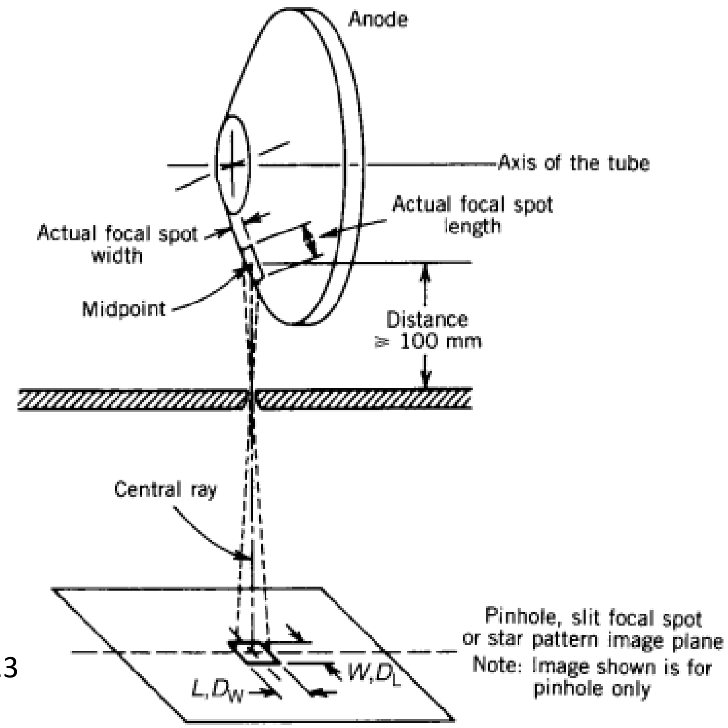
Para adaptar a intensidade do feixe à forma anatómica do paciente
(mais grosso no meio)

O tubo de raios-X: focagem

Princípio da focagem linear



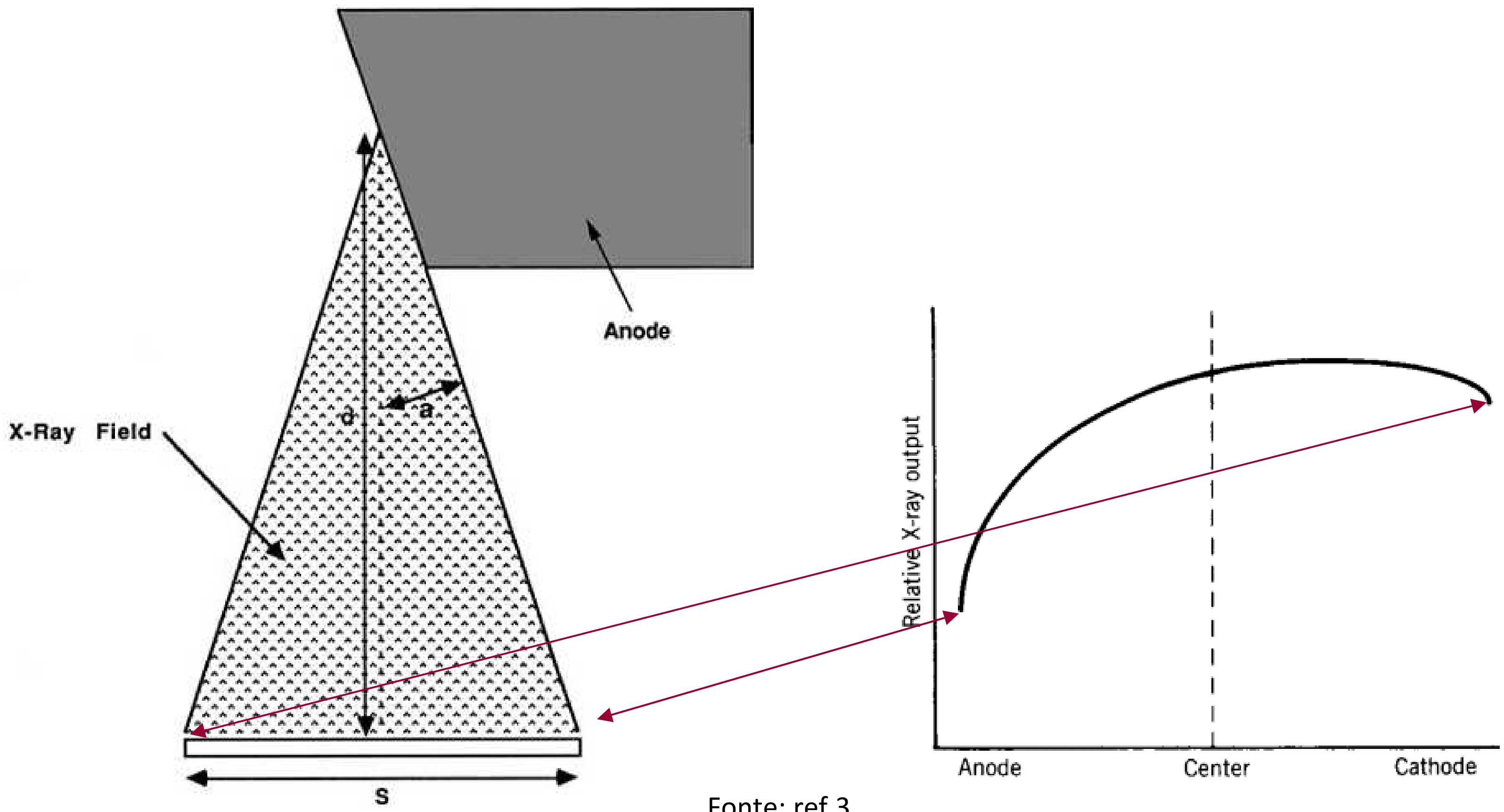
$$f = F \sin(a)$$



Fonte: ref.3

Princípio de medida do tamanho do foco usando uma câmara de "pinhole"

O tubo de raios-X: campo iluminado



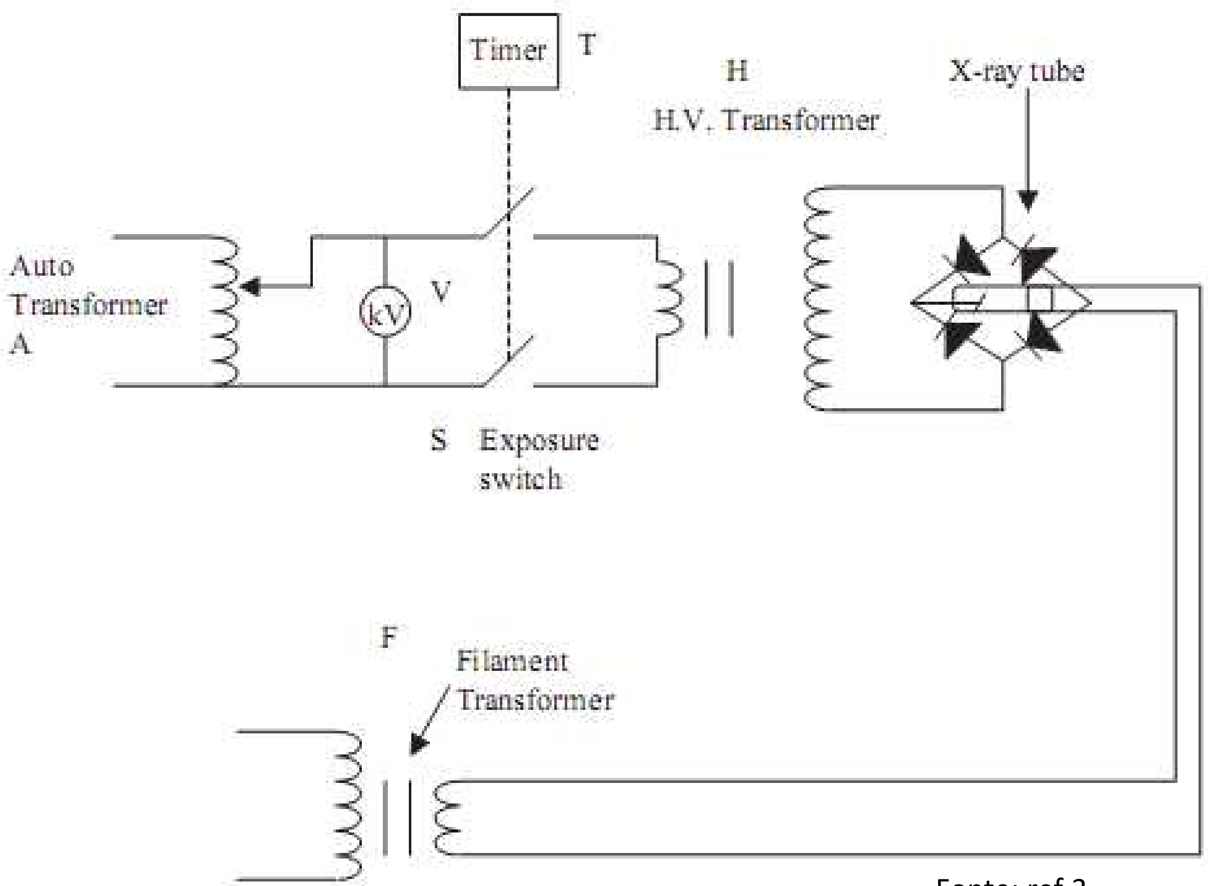
Fonte: ref.3

Tamanho de campo

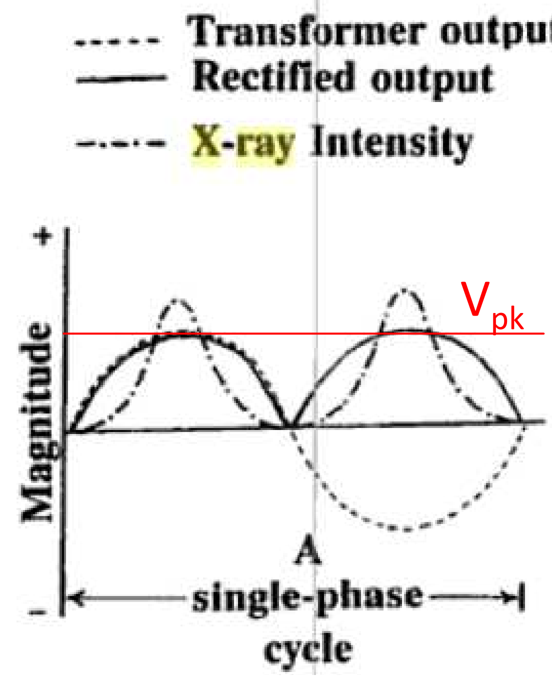
Efeito de Heel

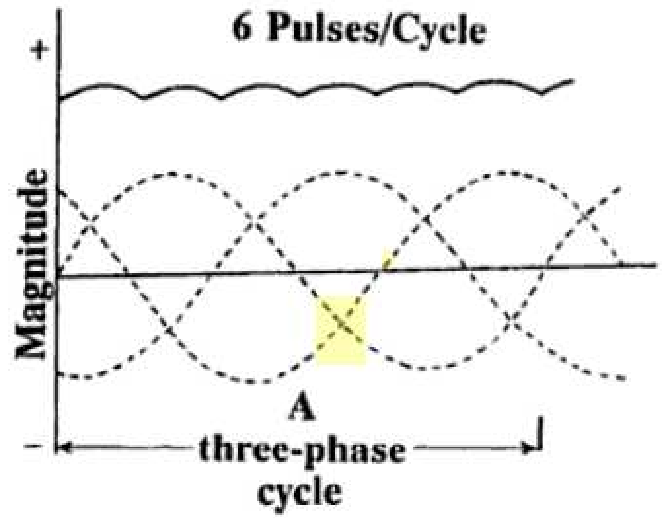
$$s = 2d\tan(a)$$

O tubo de raios-X: esquema eléctrico



Fonte: ref.3





Provavelmente obsoleto.
Mas o princípio mantém-se.

O tubo de raios-X: limite térmico de funcionamento

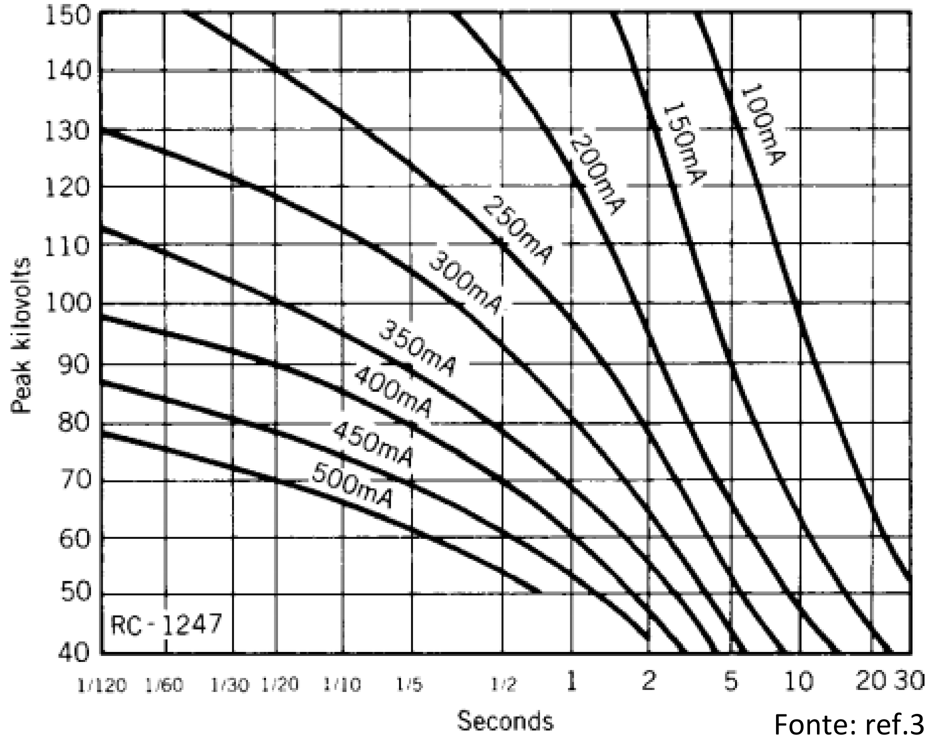
Potencia depositada no ânodo (rotativo…)

$P = K \times V_{pk} \times i$
$E = P \times t$

P = potência em Watt
E = Energia por disparo em Joule
K = factor que depende do tipo de alimentação
monofásico: 0.74
trifásico: 1
V_{pk} = tensão de pico
i = corrente eficaz
t = duração do disparo

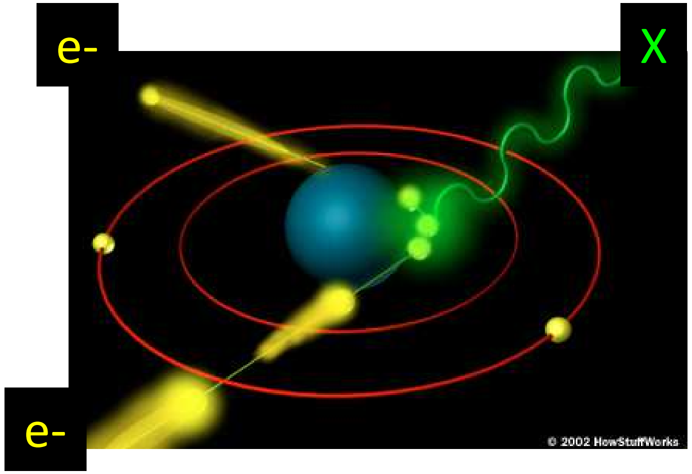
P = 37.5 kW
E = 625 J

P = 15 kW
E = 45 kJ

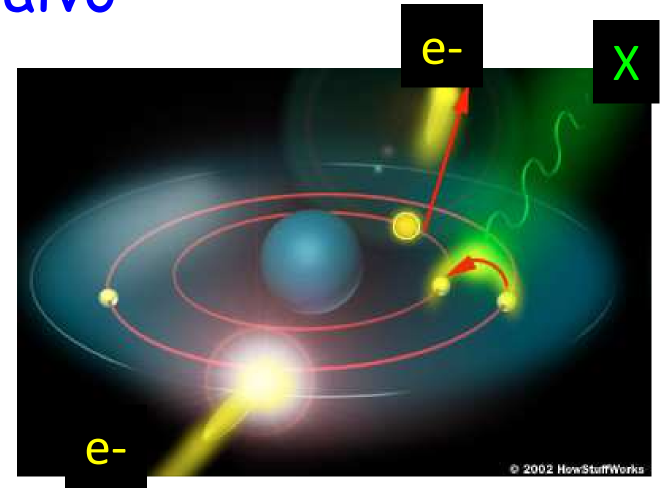


Fonte: ref.3

Interacção entre os electrões e o alvo



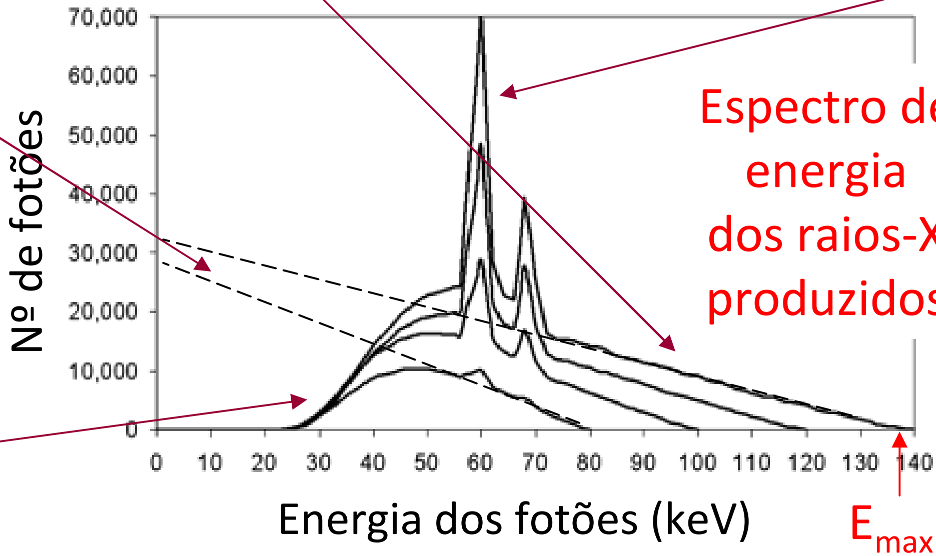
Radiação de travagem
("Bremsstrahlung ")



Radiação característica
do átomo atingido (tungsténio)

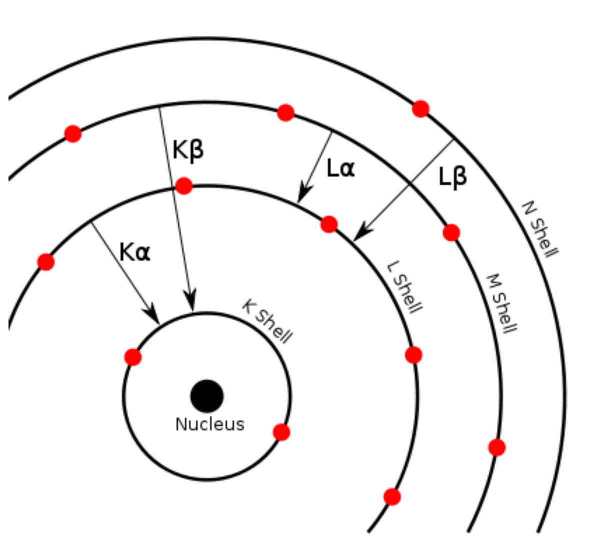
Espectro natural

Filtragem interna (alvo+janela)



Espectro de energia dos raios-X produzidos

$E_{max}=V_{pk}$



Semi-camada e energia efectiva

A semi-camada é a espessura de material necessária para reduzir a intensidade do feixe a metade do seu valor inicial. Depende do material.

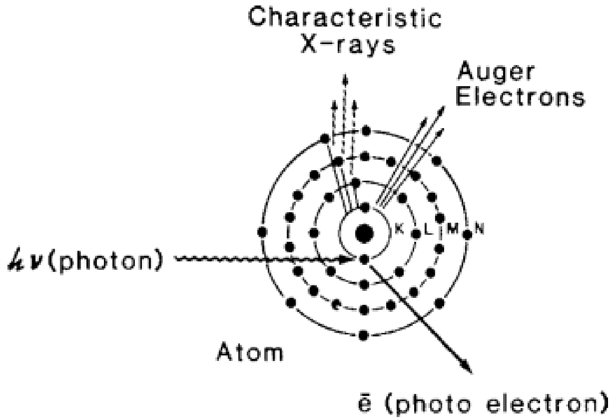
X-ray energy (keV)	HVL, muscle (cm)	HVL, bone (cm)
30	1.8	0.4
50	3.0	1.2
100	3.9	2.3
150	4.5	2.8

A qualidade espectral de um feixe pode ser expressa em termos da sua semi-camada de Al.

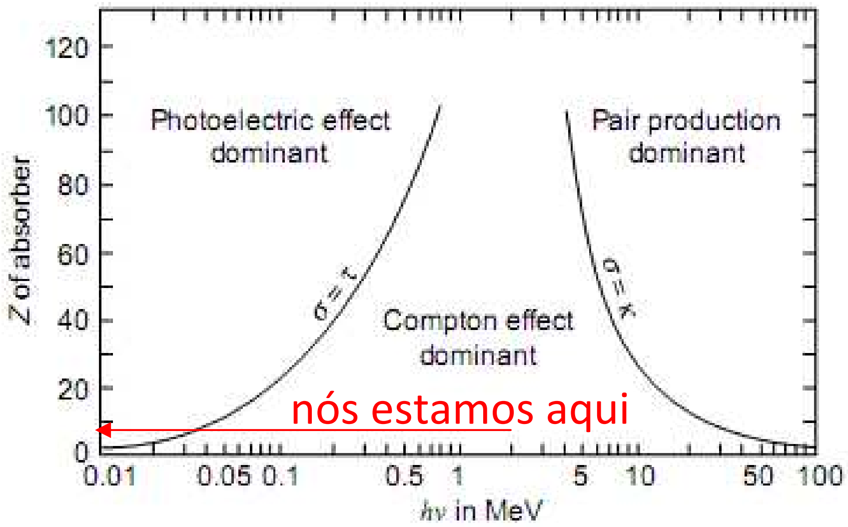
A energia efectiva de um feixe policromático é a energia de um feixe monocromático que teria a mesma semi-camada de Al. Por exemplo, para alvo de W a 150kV$_{pk}$ ~ 68keV

Interacção entre os fotões e a matéria

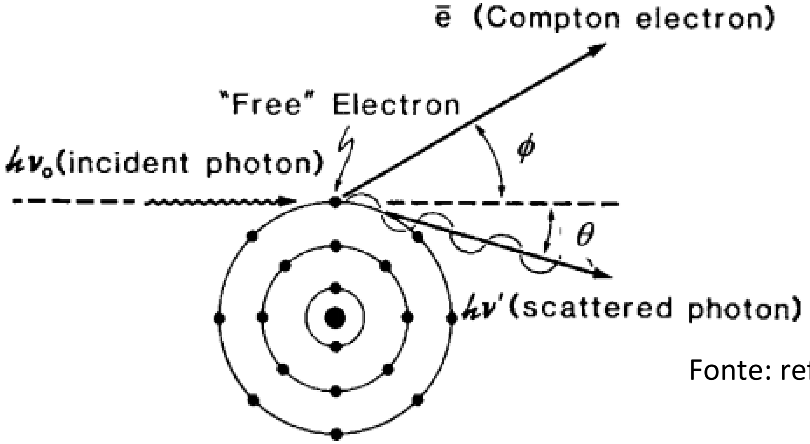
Efeito fotoeléctrico



Fonte: ref.3

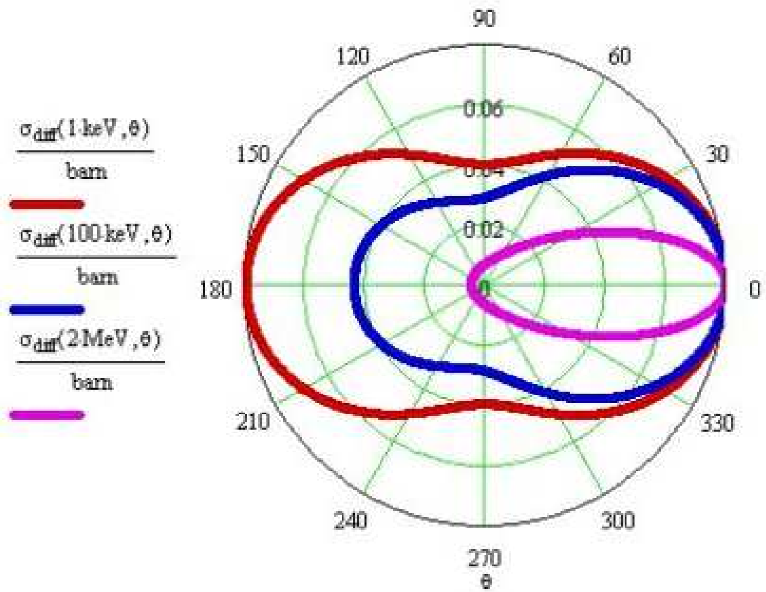


Difusão de Compton ou inelástica ou incoerente



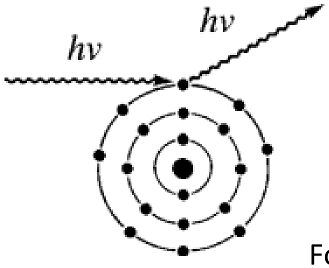
Fonte: ref.3

Distribuição angular do fotão difundido



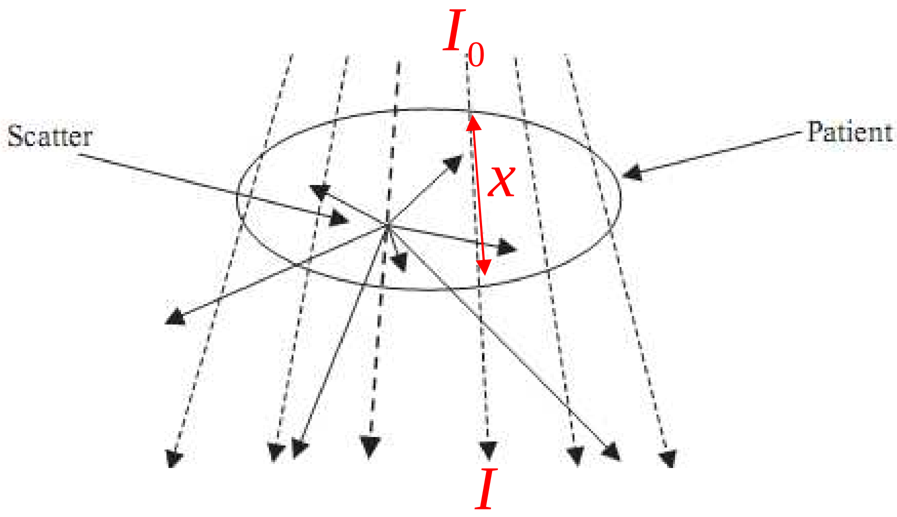
Interacção entre os fotões e a matéria

Difusão de Rayleigh ou elástica ou coerente
(pouco importante)



Fonte: ref.3

Penetração dos fotões no organismo

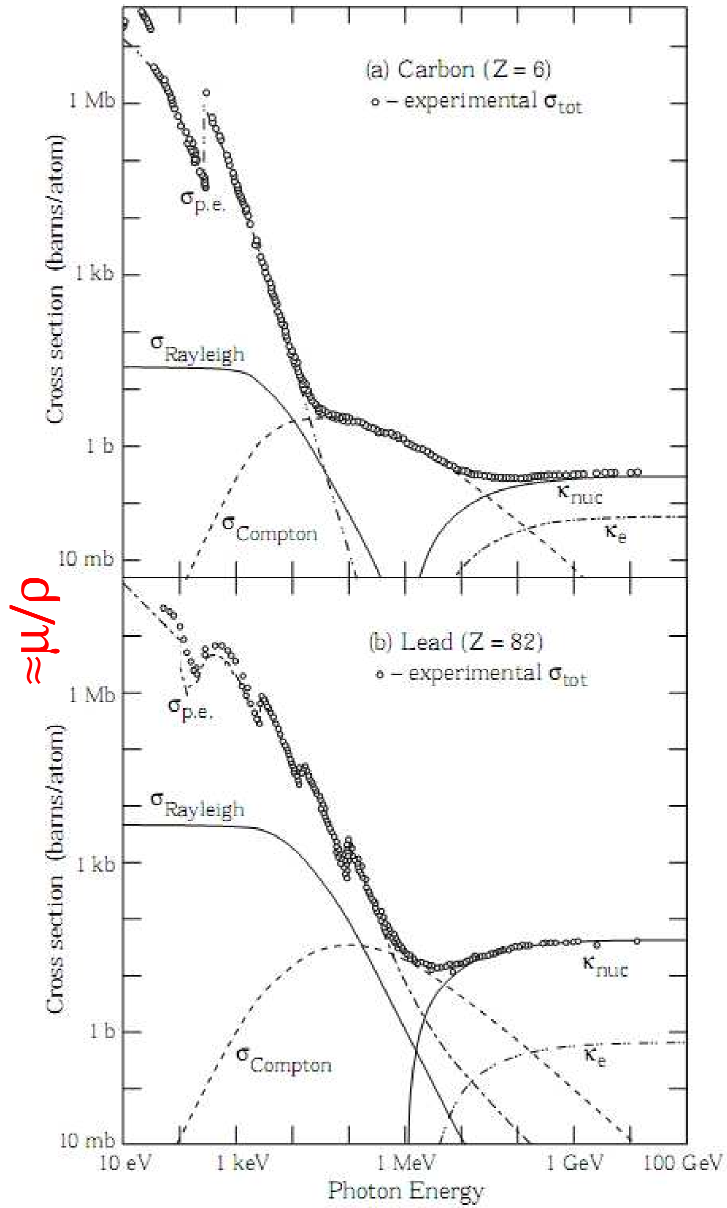


$I / I_0 = e^{-\mu x} = e^{-(\mu/\rho)(\rho x)}$ (para cada raio)

$\mu = \mu_{fotoeléctrico} + \mu_{Compton} + \mu_{Rayleigh}$

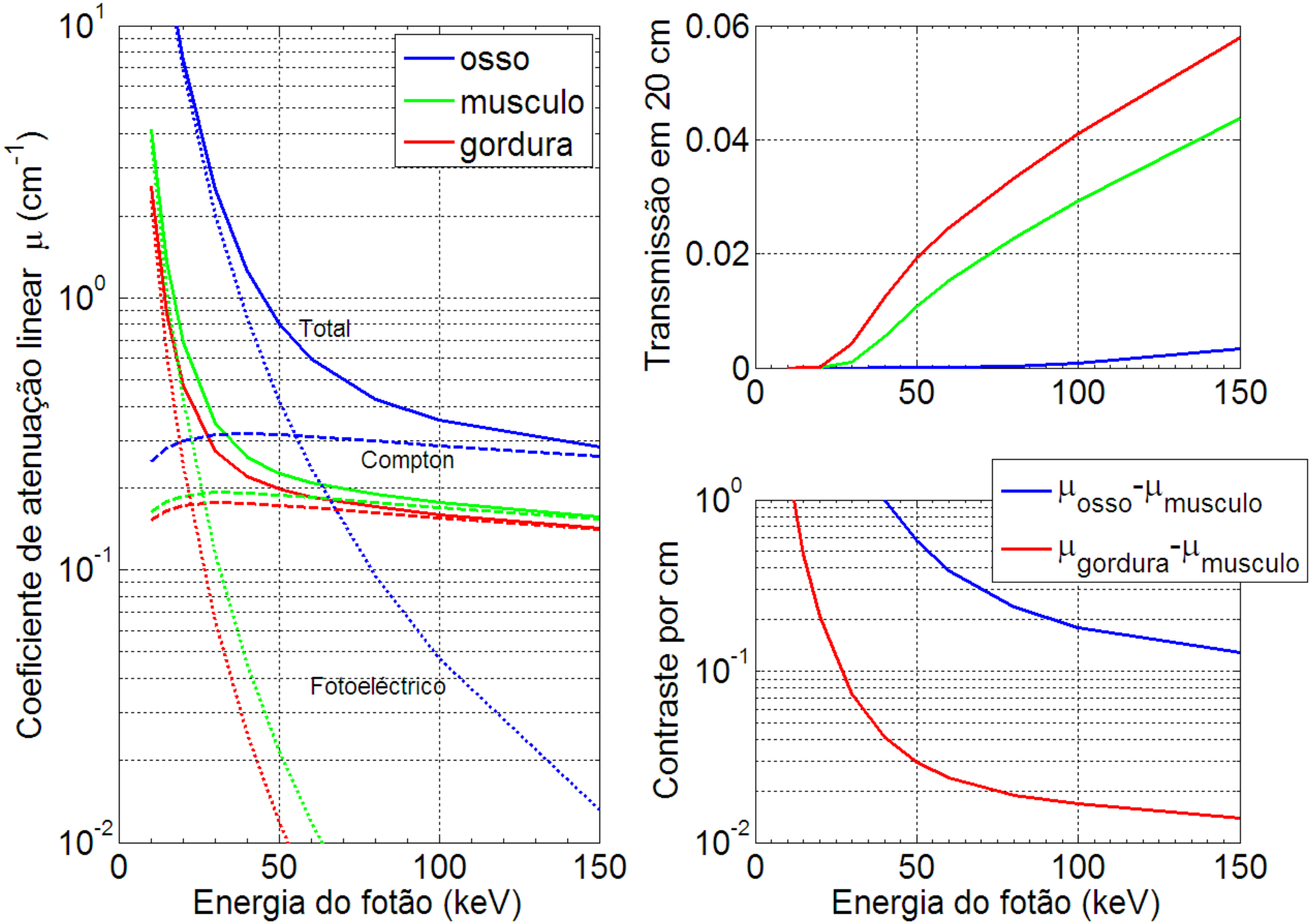
μ = coeficiente de atenuação linear [cm^{-1}]
x = espessura atravessada [cm]
μ/ρ = coeficiente de atenuação mássica [cm^2/g]
ρx = espessura mássica [g/cm^2]

Nota importante: I é a intensidade do feixe de fotões que não interagiram, não é o mesmo que a intensidade do feixe que emerge do paciente, pois muitos fotões são difundidos e não absorvidos



Formação da imagem em raios-X

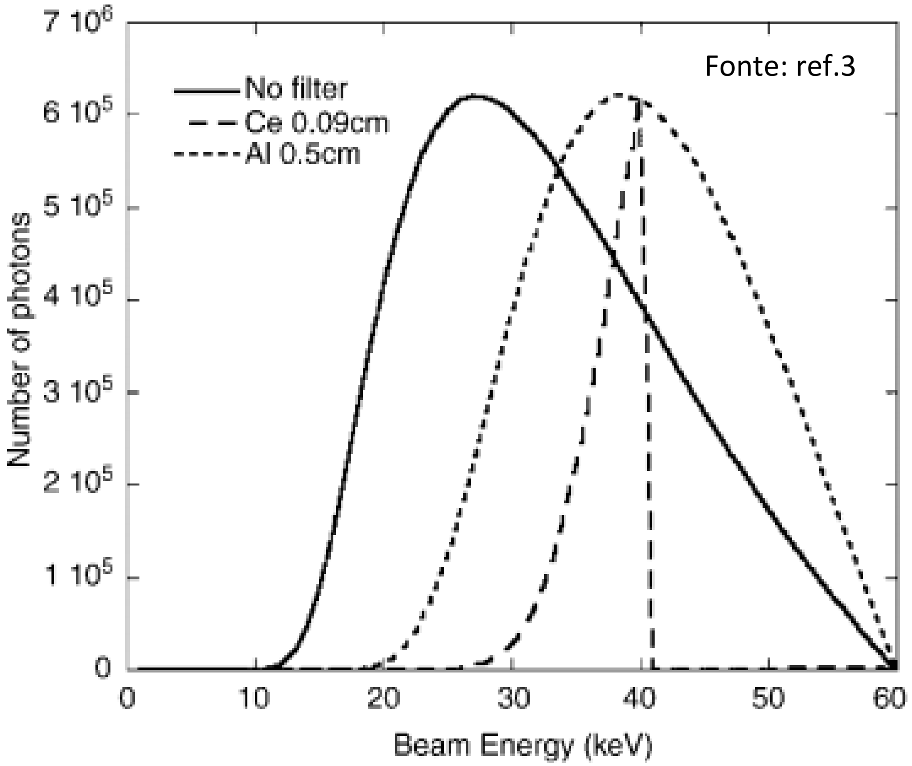
Transmissão vs. contraste



Dados obtidos em
http://physics.nist.gov/PhysRefData/XrayMassCoef/cover.html
http://physics.nist.gov/PhysRefData/Xcom/Text/XCOM.html

Filtragem

Os fotões de baixa energia são absorvidos totalmente no paciente e logo não contribuem para a imagem, mas contribuem para a dose absorvida. Assim, estes fotões devem ser removidos do feixe através da interposição de filtros “endurecedores”.



Filtragem padrão

< 50 kV_{pk}: 0.5 mm Al

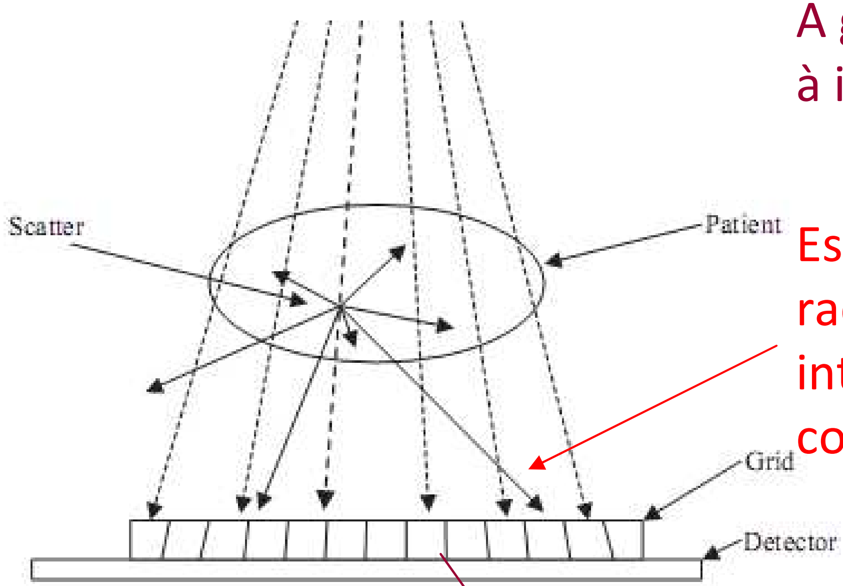
50 a 70 kV_{pk}: 1.5 mm Al

> 70 kV_{pk}: 2.5 mm Al

Detecção dos raios-X em radiografia projectiva

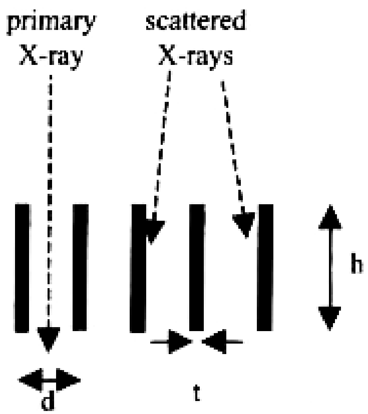
Grelha anti-difusão

A grelha fixa a distancia fonte-imagem devido à inclinação das suas paredes



Estes fotões (80% a 90% do total numa radiografia torácica) estão “fora do sítio” e introduzem um fundo difuso, reduzindo o contraste da imagem

Fonte: ref.1



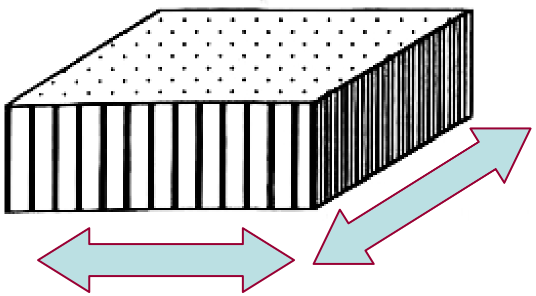
Características típicas

$$4 \le d / t \le 16$$

$$25 \le \frac{1}{d+t} \le 60\, cm^{-1}$$

$$F = I_{grelha} / I_{sem\ grelha}$$

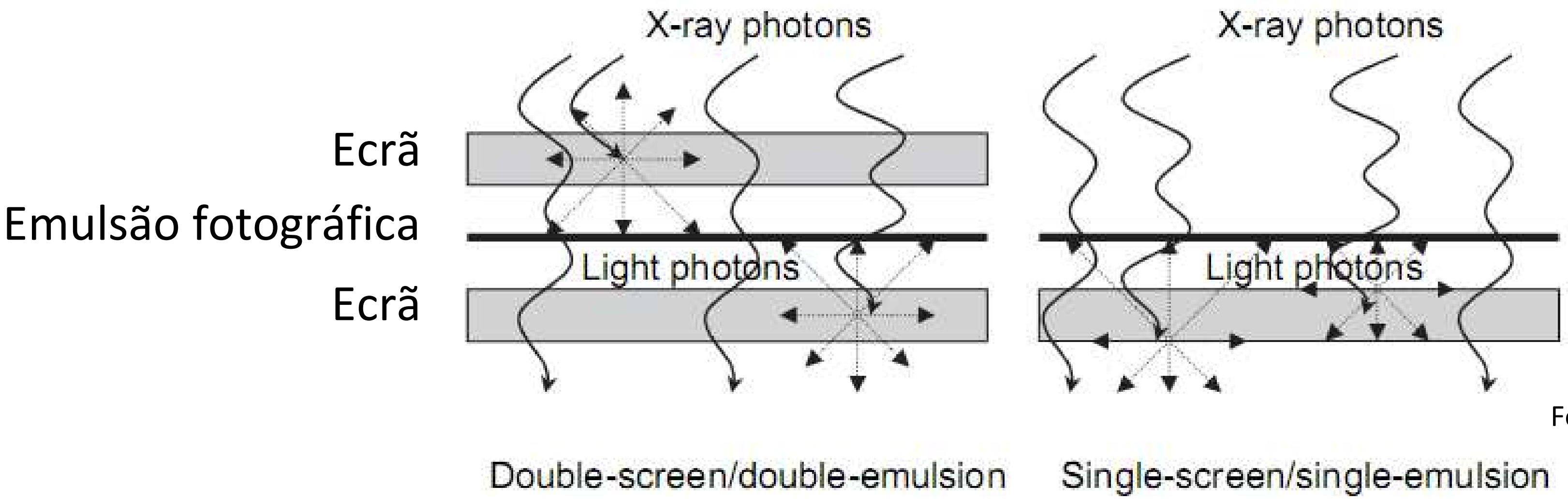
$$F = \text{"Bucky factor"}$$



Movimento oscilante

Ecrã intensificador

Constituído por compostos cintiladores de nº atómico elevado, emite luz quando absorve um raio-X, o que aumenta muito a sensibilidade aos raios-X do detector.



Fonte: ref.3

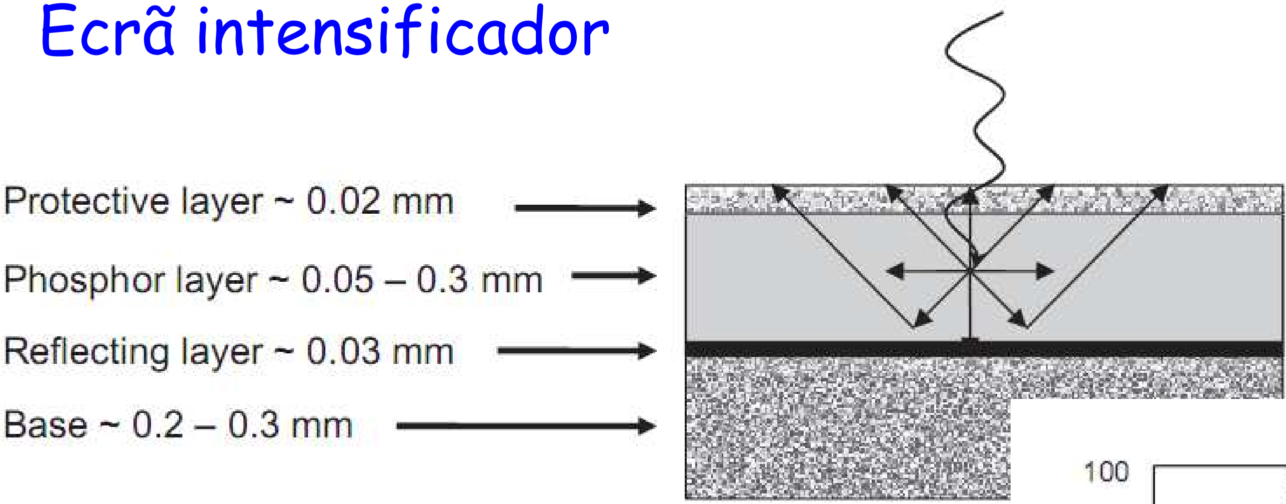
Table 1. Physical Properties of Some Common Phosphors[a]

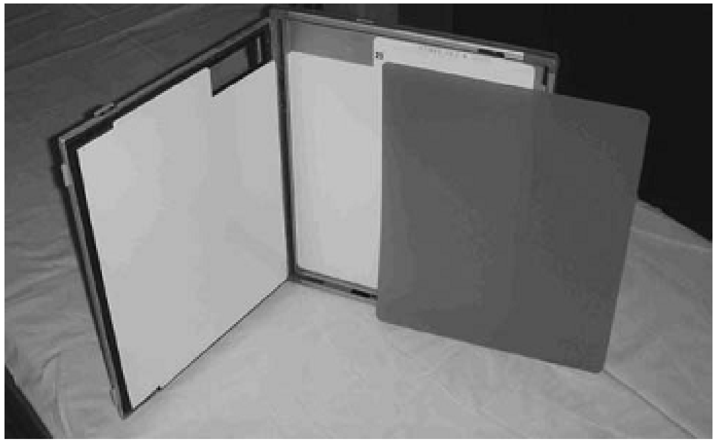
Phosphor	Atomic Number of Heaviest Element	K-Edge, keV	Conversion Efficiency, %	Light Emission Spectrum
Calcium tungstate, $CaWO_4$	74	69.5	3.5	Blue (340–540 nm)
Barium strontium sulfate, $BaSO_4$:Eu	56	37.4	6	Blue (330–430 nm)
Barium fluorochloride, BaFCl:Eu	56	37.4	13	Blue (350–450 nm)
Gadolinium oxysulfide, Gd_2O_2S:Tb	64	50.2	15	Green (400–650 nm)
Lanthanum oxybromide, LaOBr	57	38.9	13	Blue (360–620 nm)
Lanthanum oxysulfide, La_2O_2S:Tb	57	38.9	12	Green (480–650 nm)
Yttrium oxysulfide, Y_2O_2S:Tb	39	17.0	18	Blue (370–630 nm)

Detecção dos raios-X em radiografia projectiva

Ecrã intensificador

Fonte: ref.3





"cassete" radiográfica

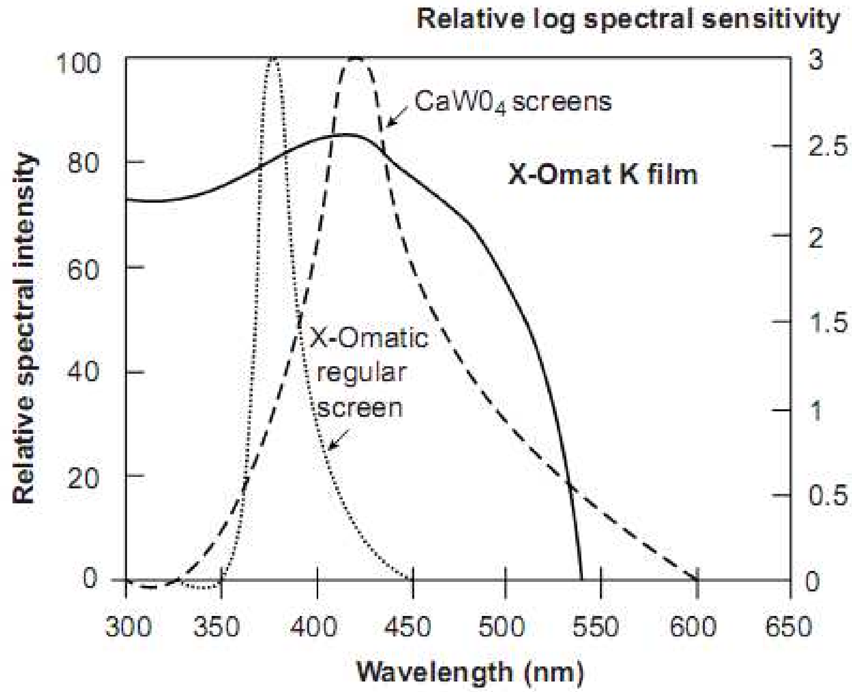


Figure 9. Relative spectral match of Kodak X-Omat K film with ultraviolet (UV) and blue-emitting screens. (Courtesy of Dr. R. Dickerson of Eastman Kodak Company.)

Detecção dos raios-X em radiografia projectiva

Emulsão fotográfica

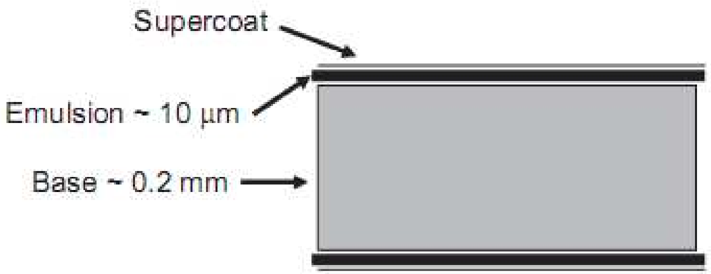
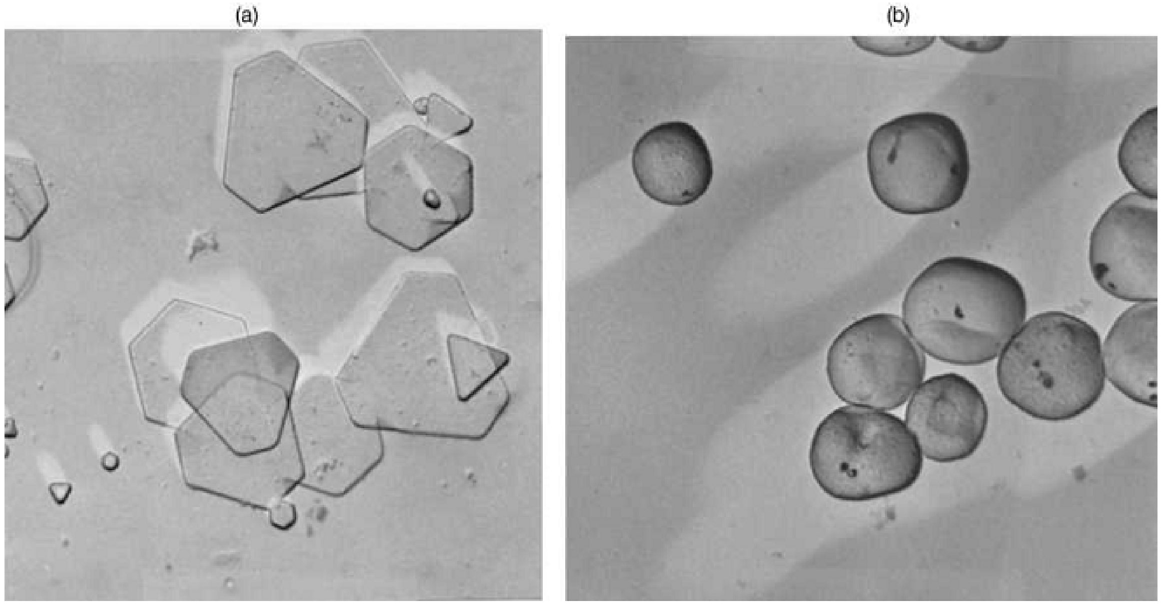


Figure 4. A cross-sectional view of an X-ray film.

A probabilidade dos fotões interagirem aqui é quase nula, daí o ecrã intensificador

Fonte: ref.3

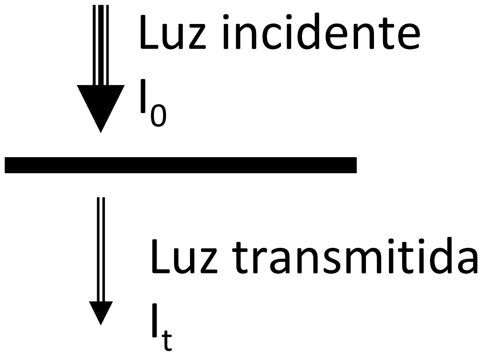


Cristais de halogeneto de prata (AgBr + AgCl ou AgI) em emulsão (gelatina)

Detecção dos raios-X em radiografia projectiva

Emulsão fotográfica

Densidade óptica D



$$D = \log\left(\frac{I_0}{I_t}\right)$$

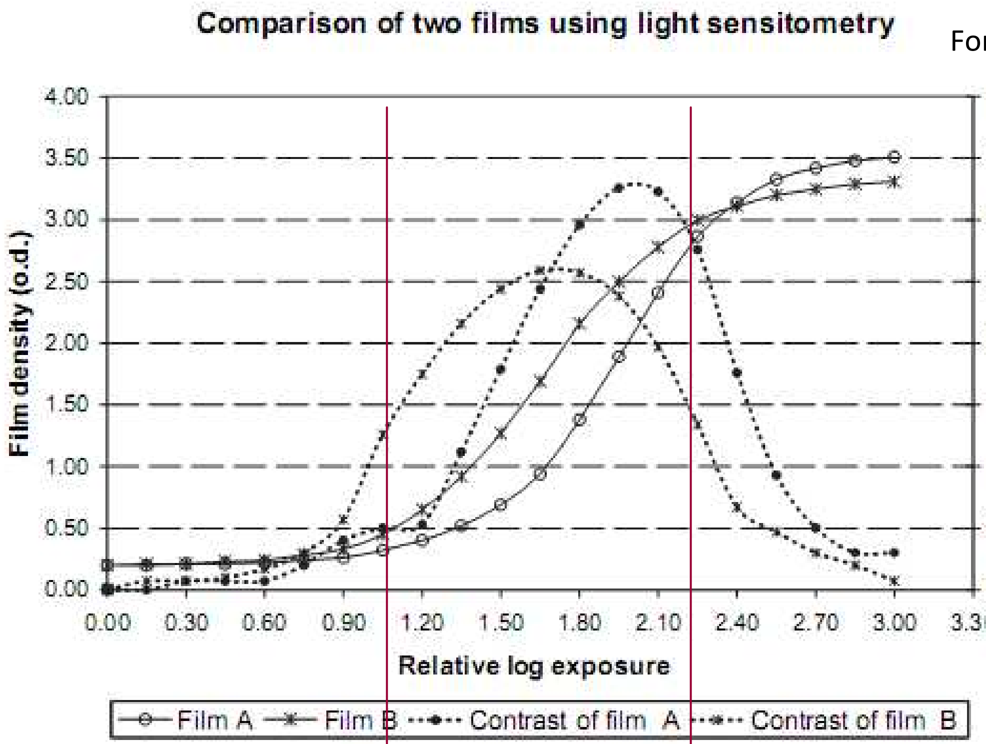
Corresponde à sensação visual de luminosidade

Contraste

$$\gamma = \left.\frac{D_2 - D_1}{\log(E_2) - \log(E_1)}\right)_{\max}$$

E = Exposição

Fonte: ref.3



Região útil do filme *

A relação entre exposição e D é muito não-linear excepto num intervalo estreito de exposições. Fora deste intervalo, o contraste é baixo. Este é um grave inconveniente técnico da emulsão.

Detecção dos raios-X em radiografia projectiva

Radiografia digital = detector electrónico

Fonte: ref.4

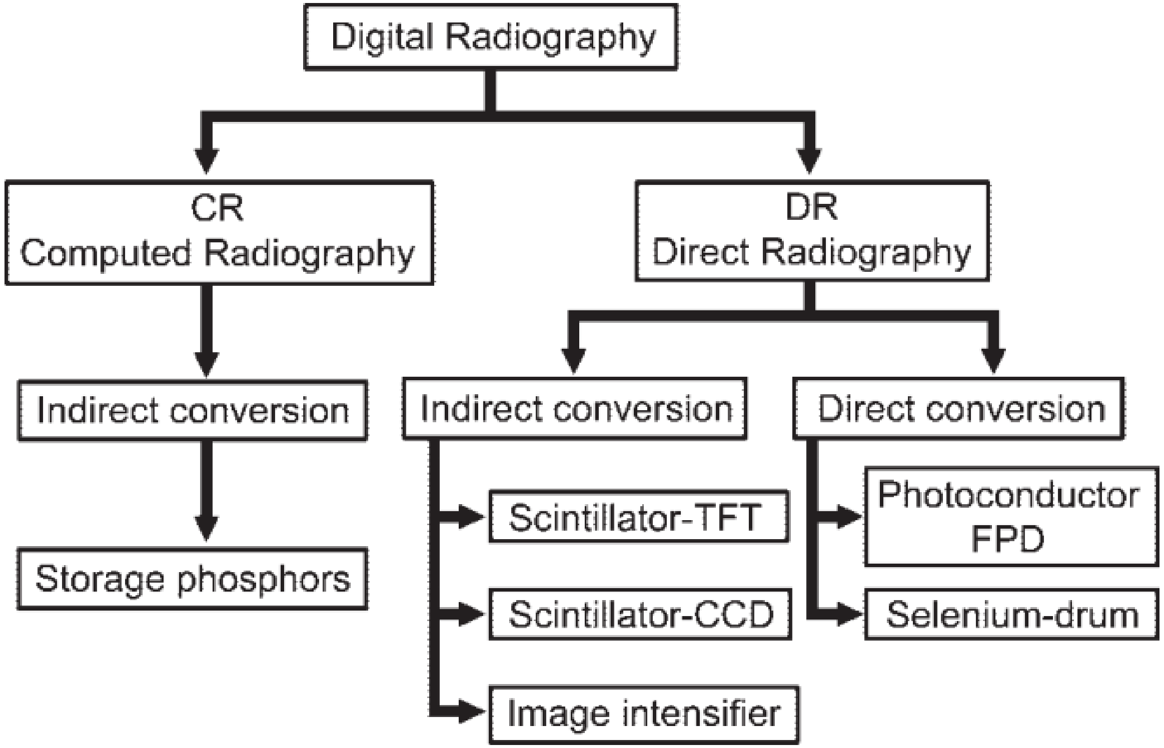


Figure 2. Chart provides a systematic overview of various types of digital detectors. *CCD* = charge-coupled device, *FPD* = flat-panel detector, *TFT* = thin-film transistor.

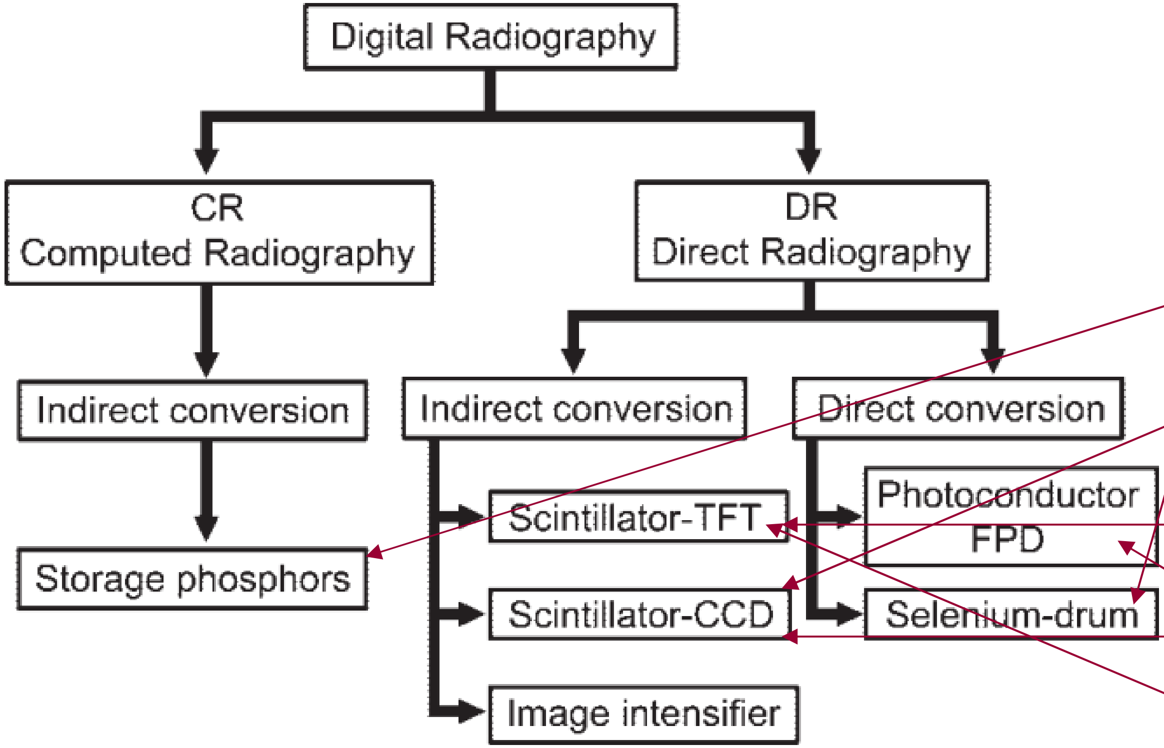
Table 1
Timetable of Developments in Digital Radiography

Year	Development
1977	Digital subtraction angiography
1980	Computed radiography (CR), storage phosphors
1987	Amorphous selenium–based image plates
1990	Charge-coupled device (CCD) slot-scan direct radiography (DR)
1994	Selenium drum DR
1995	Amorphous silicon–cesium iodide (scintillator) flat-panel detector
1995	Selenium-based flat-panel detector
1997	Gadolinium-based (scintillator) flat-panel detector
2001	Gadolinium-based (scintillator) portable flat-panel detector
2001	Dynamic flat-panel detector fluoroscopy–digital subtraction angiography

Detecção dos raios-X em radiografia projectiva

Radiografia digital = detector electrónico

Fonte: ref.4



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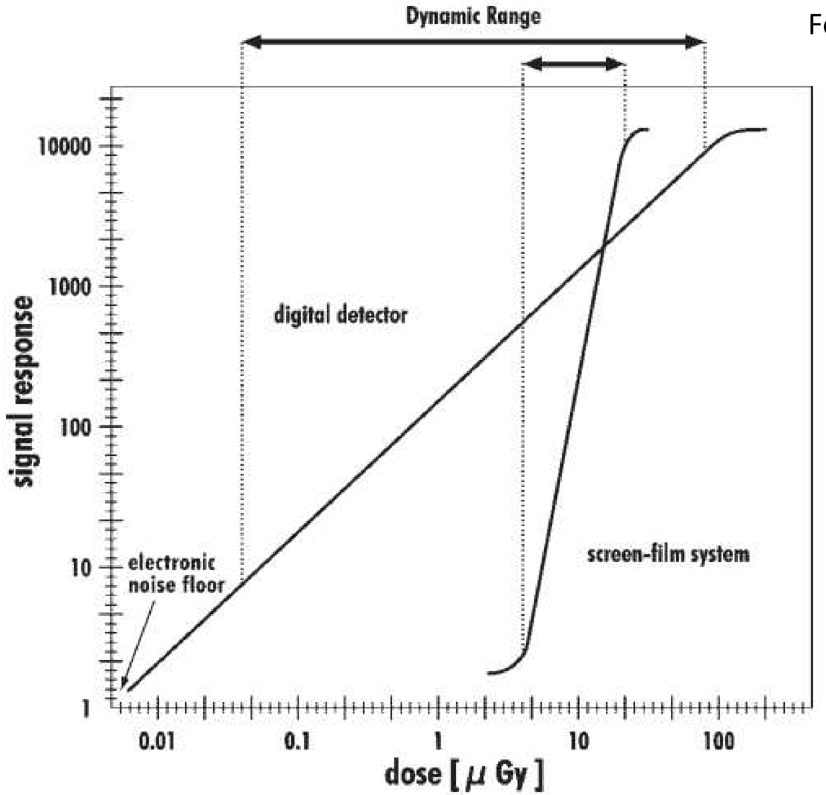
Radiografia digital = detector electrónico



Fonte: ref.4

Figure 8. Graph illustrates the dynamic range of screen-film combinations and digital detectors. Screen-film systems have only a limited tolerance for radiation exposure, resulting in a steep and tight curve, whereas the curve for digital detectors is less steep and covers a wider range. As a result, an optimal signal response will occur over a wider exposure range with digital detectors than with screen-film combinations.

Vantagens:
- extraordinária linearidade
- suporte digital
- tratamento de imagem

Desvantagens
- caro
- suporte digital
- alguns sistemas são fixos

Computed radiography (CR) /image plate

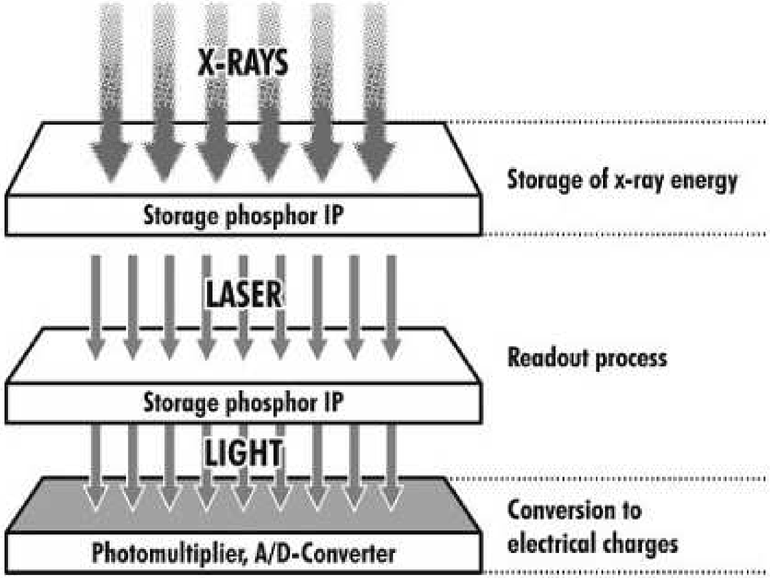
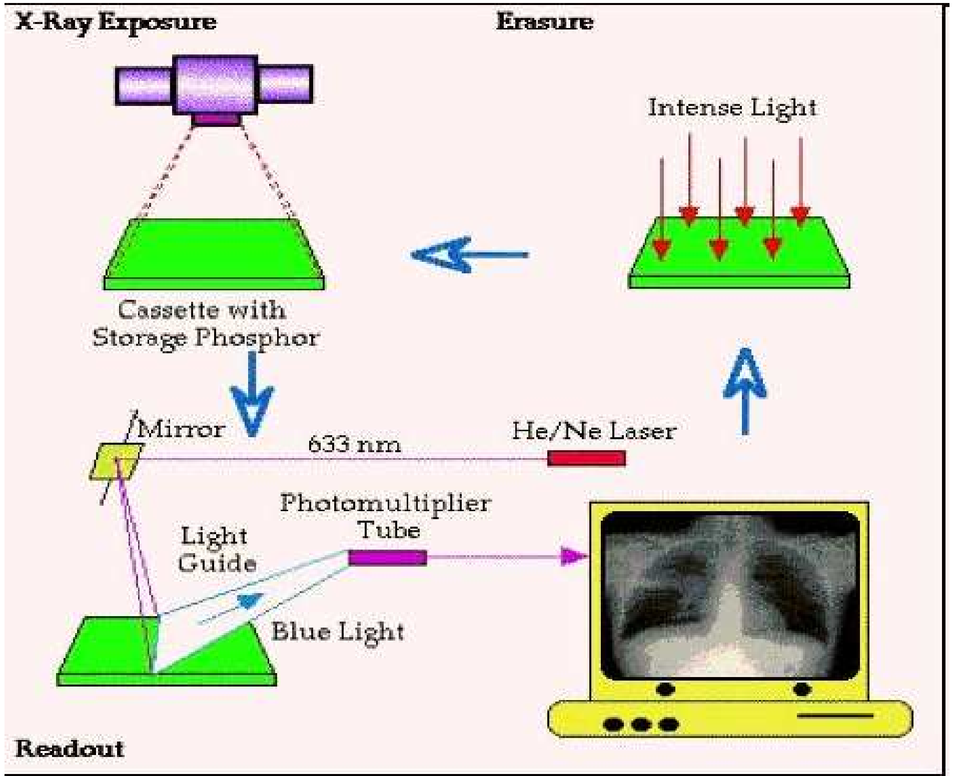
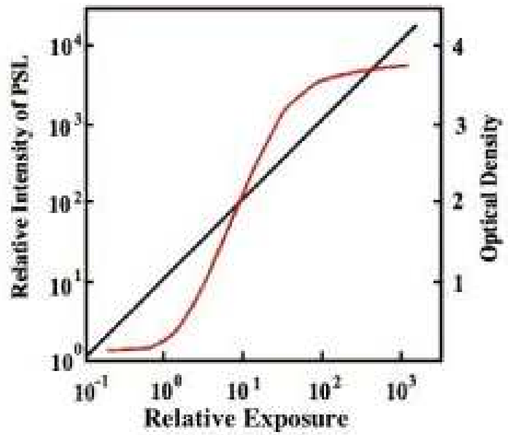


Figure 3. Drawing illustrates a CR system based on storage-phosphor image plates. Image generation is separated into two steps. First, the image plate *(IP)* is exposed to x-ray energy, part of which is stored within the detective layer of the plate. Second, the image plate is scanned with a laser beam, so that the stored energy is set free and light is emitted. An array of photomultipliers collects the light, which is converted into electrical charges by an analog-to-digital *(A/D)* converter.

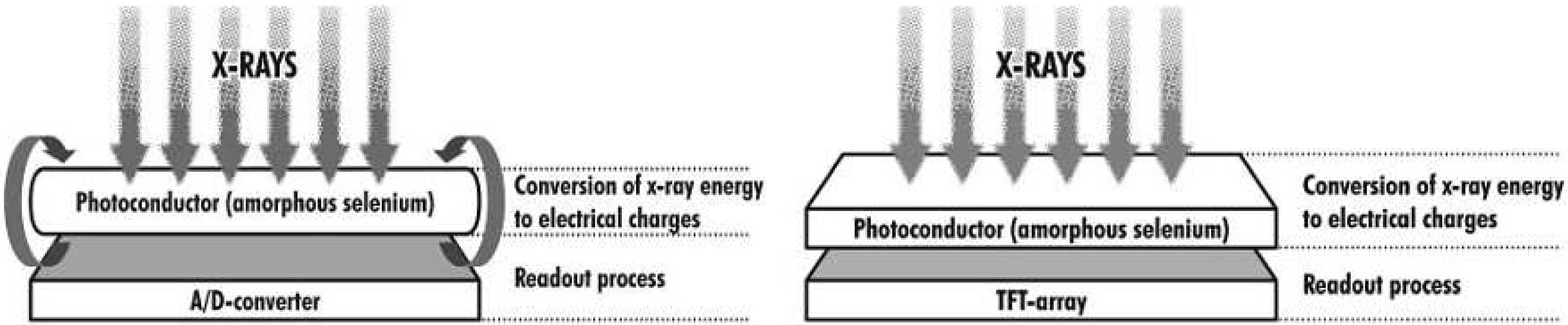
Fonte: ref.4





Fonte: ref.5

Conversão directa: selénio + electronica



a. **b.**

Figure 4. Amorphous selenium–based direct conversion DR systems. **(a)** Drawing illustrates a selenium drum–based system. A rotating selenium-dotted drum with a positive electrical surface charge is exposed to x-rays. Alteration of the charge pattern of the drum surface is proportional to the incident x-rays. The charge pattern is then converted into a digital image by an analog-to-digital *(A/D)* converter. **(b)** Drawing illustrates a selenium-based flat-panel detector system. Incident x-ray energy is directly converted into electrical charges within the fixed photoconductor layer and read out by a linked TFT array beneath the detective layer.

Fonte: ref.4

Conversão indirecta: cintilador + fotodetector CCD

Fonte: ref.4

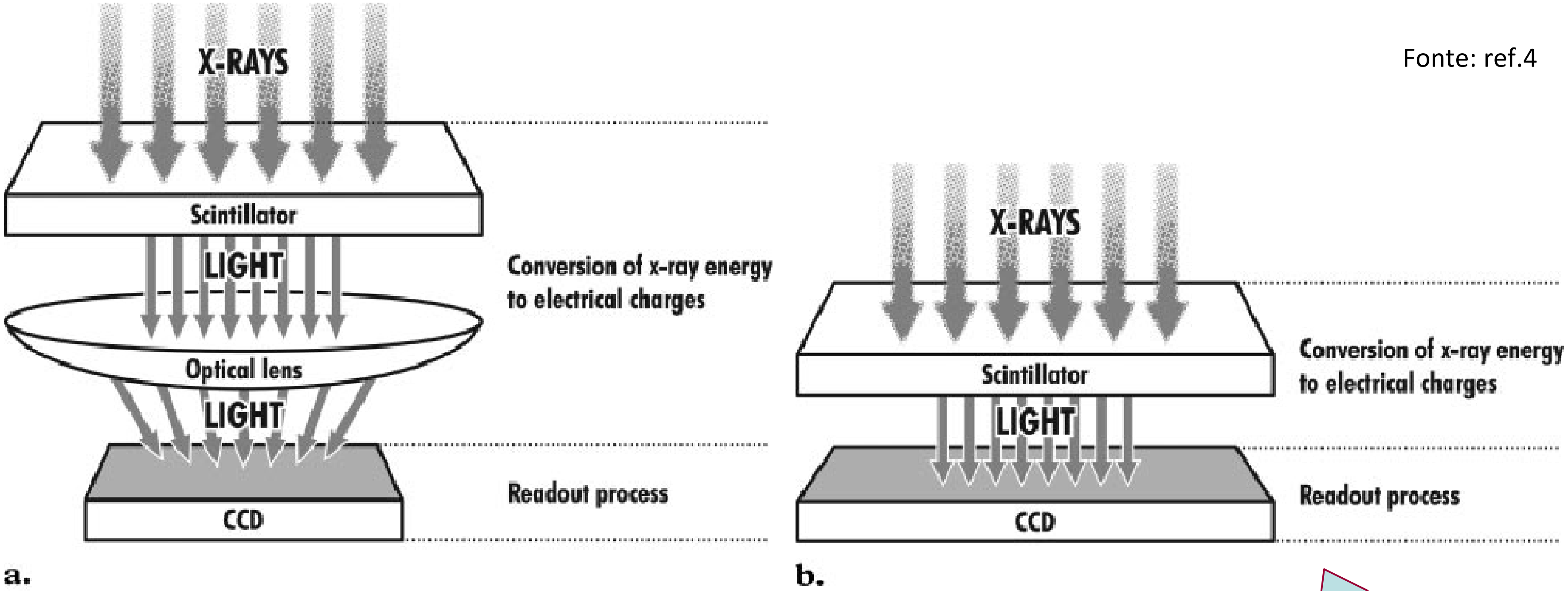


Figure 5. CCD-based indirect conversion DR system. **(a)** Drawing illustrates a lens-coupled CCD-based system. The incident x-ray energy is converted into light by a scintillator. The emitted light has to be bundled by an optical lens to fit the size of the CCD chip, which subsequently converts the light energy into electrical charges. **(b)** Drawing illustrates a slot-scan CCD-based system. The patient is scanned with a fan-shaped beam of x-rays. A simultaneously moving CCD detector of the same size collects the emitted light and converts the light energy into electrical charges.

Varrimento

Conversão indirecta: cintilador + matriz de fotodiodos+TFT

Fonte: ref.4

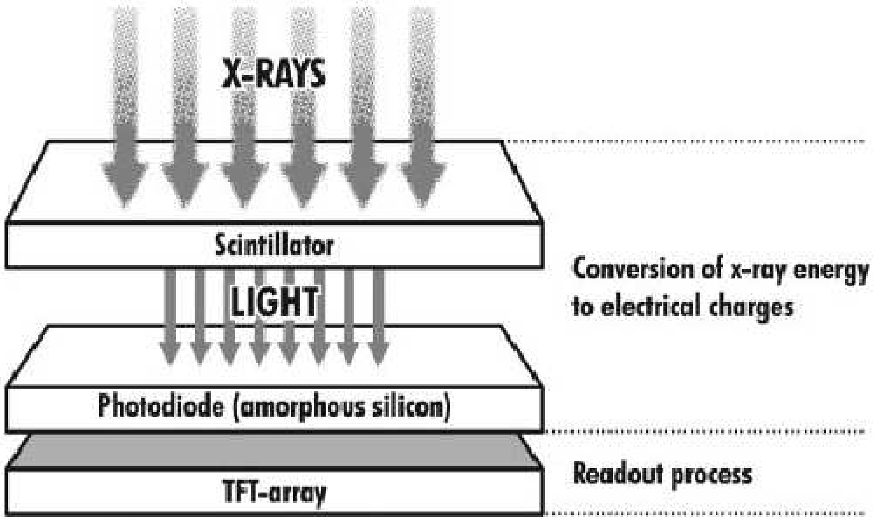


Figure 6. Drawing illustrates an amorphous silicon–based indirect conversion DR system. X-ray energy is converted into visible light in a scintillator layer. The emitted light is then converted into electrical charges by an array of silicon-based photodiodes and read out by a TFT array.

Comparação entre os diversos sistemas

Fonte: ref.4

Table 2
Technical Features of Various Digital Radiography Systems

Feature	Type of System						
	Screen-Film	Storage-Phosphor	Lens-coupled CCD	Slot-Scan CCD	Direct FPD	Indirect FPD	Indirect FPD
Converter	Gd_2O_2S	BaSrFBr:Eu	Gd_2O_2S	CsI:TI	Selenium	Gd_2O_2S	CsI:TI
Readout	Film	Laser	CCD	CCD	Active selenium matrix	Active silicon matrix	Active silicon matrix
Detector size (in)	14 × 17	14 × 17	14 × 17	17 × 17	14 × 17	17 × 17	17 × 17
Pixel size (μm)	. . .	200	167	162	139	160	143
Matrix	. . .	1760 × 2140	2000 × 2500	2736 × 2736	2560 × 3072	2688 × 2688	3121 × 3121
Nyquist frequency (cycles/mm)	5	2.5	3.0	3.1	3.6	3.1	3.5
Dynamic range	1:30	1:40,000	>1:4000	1:10,000	>1:10,000	>1:10,000	>1:10,000

Note.—FDP = flat-panel detector. =TFT

Comparação entre os diversos sistemas - DQE

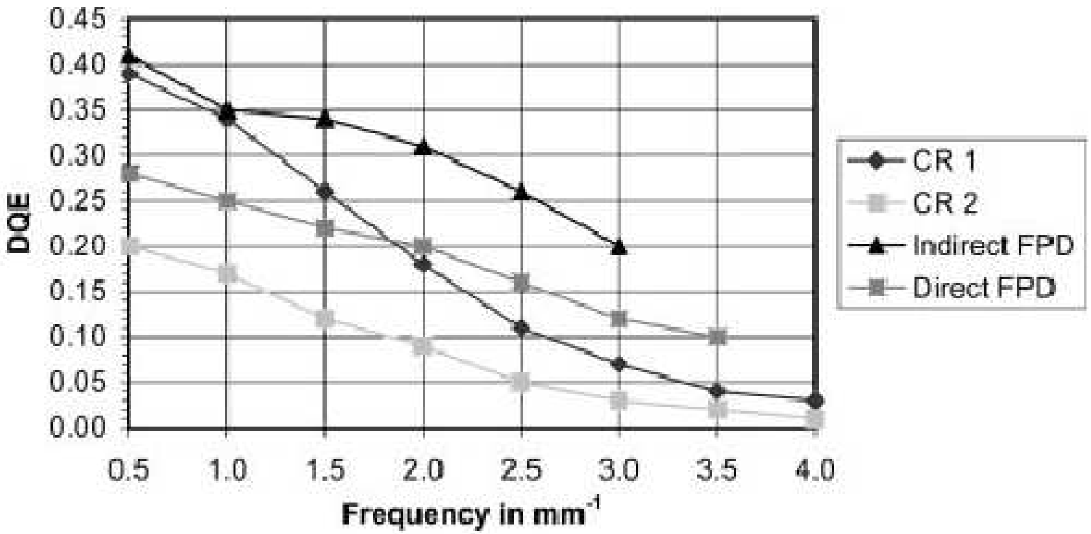


Figure 9. Graph illustrates the DQE curves for four digital detectors. *CR 1* = needle-structured storage phosphor and line scanner (MD5.0/DX-S; Agfa-Gevaert, Mortsel, Belgium), *CR 2* = unstructured storage phosphor and flying-spot scanner (MD40/ADC Compact, Agfa-Gevaert), *Indirect FPD* = CsI-based flat-panel detector (Pixium 4600; Trixell, Moirans, France), *Direct FPD* = selenium-based flat-panel detector (DR 9000; Kodak, Rochester, NY). Fonte: ref.4

DQE=Detective Quantum Eficiency

$$DQE = \left[\frac{SNR_{out}}{SNR_{in}}\right]^2$$

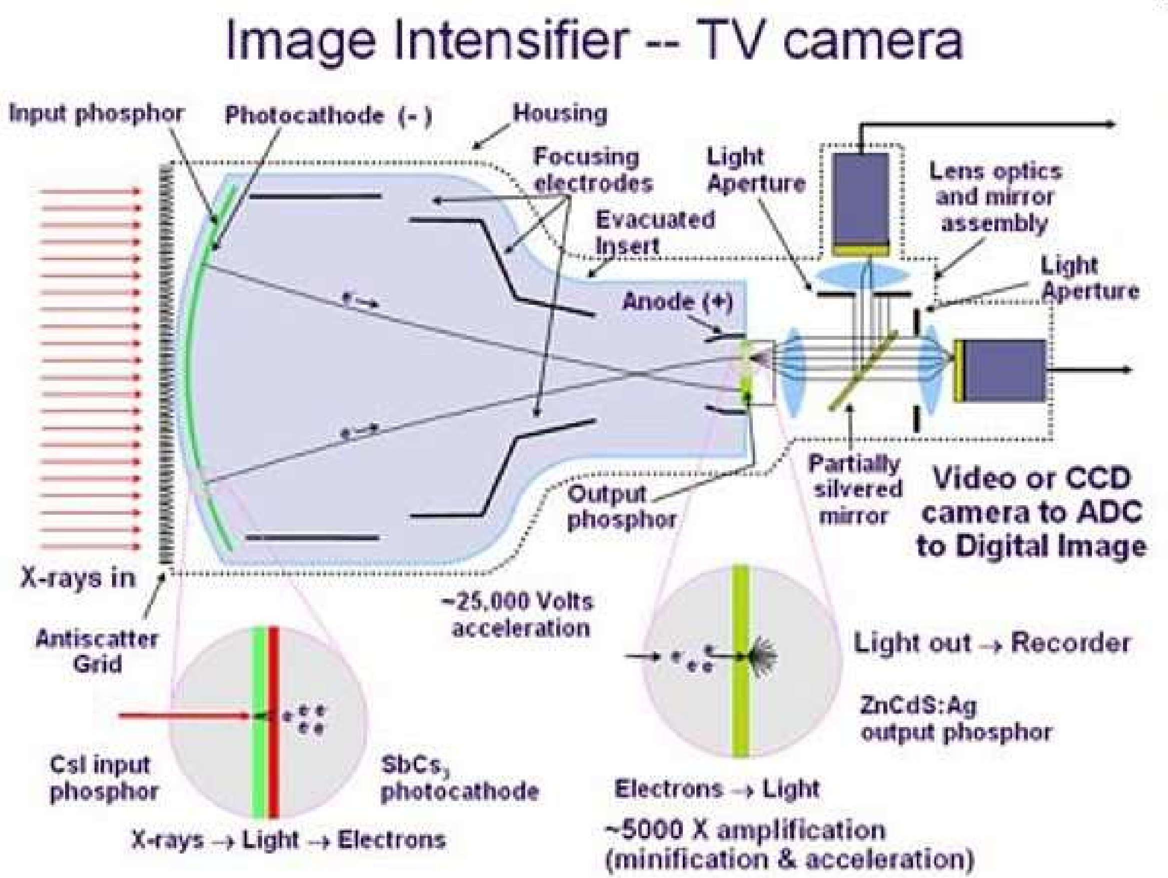
$$SNR = \text{Signal to Noise Ratio}$$

$$SNR = RMS(\sin al)/RMS(ruido)$$

RMS=Root Mean Square

$$\text{RMS}=\sqrt{\sum (\text{v}_\text{i}\text{-<v>})^2 / N}$$

Fluoroscopia



Dose elevada para o paciente.

Origem das figuras

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Energia dupla