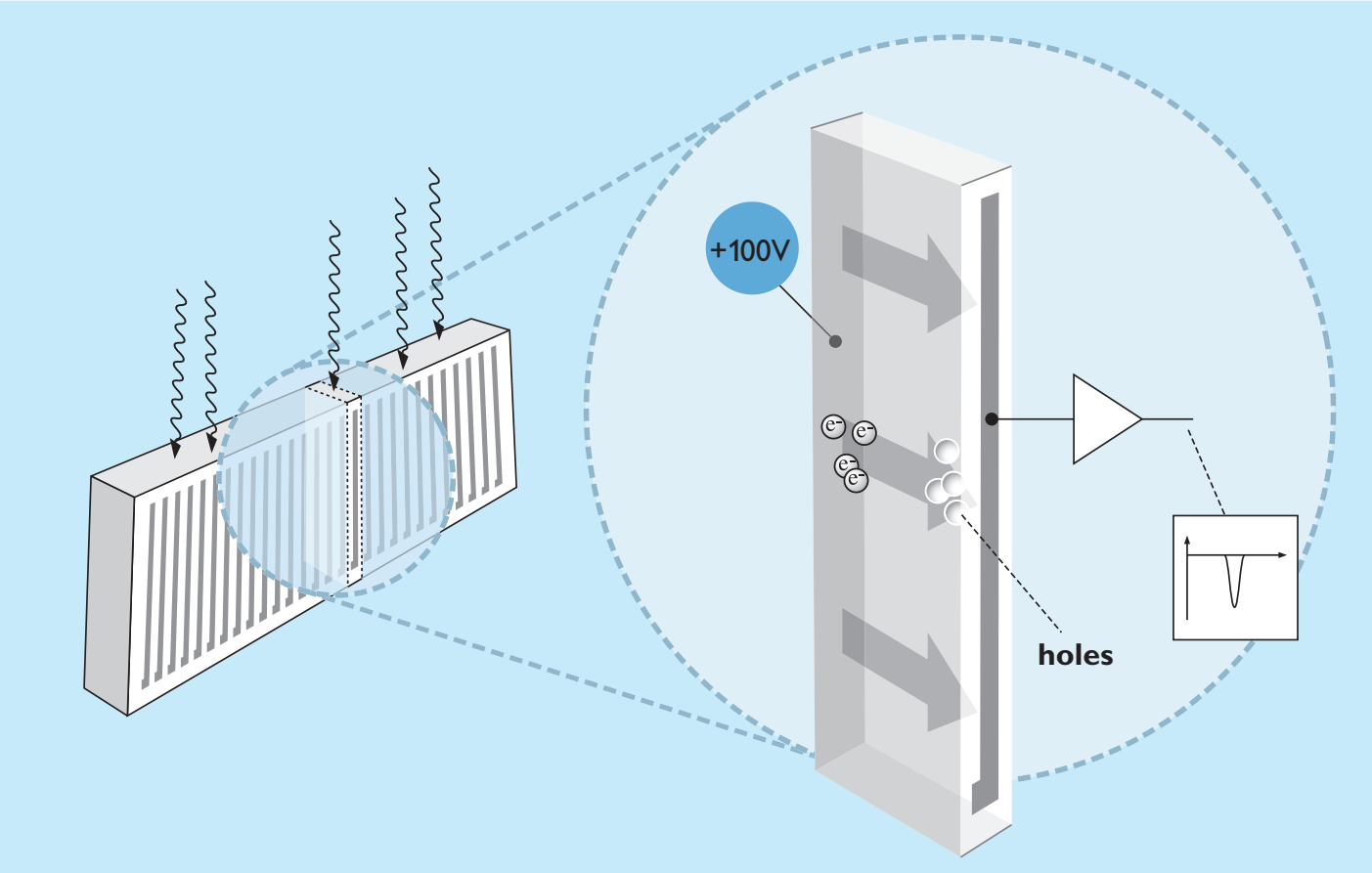


Philips MicroDose

Photon counting technology



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Requirements for detectors

Of all medical X-ray applications, mammography puts the highest demands on the imaging system. As a result, mammography is the last X-ray-based imaging technology to convert from film-screen to digital detector techniques.

For the image quality to be optimized and the radiation dose to be reduced, all the links in the imaging chain have to be considered; no imaging system is stronger than its weakest link. For a digital mammography system, the following requirements have to be met:

High image quality

- High-spatial resolution is needed to visualize and characterize the small structures in breast tissue. The detector element size and spacing define the maximum spatial resolution in an image. In mammography, a spatial resolution of 50 μm to 100 μm is required. This results in a limiting spatial resolution of 10 line pairs per mm to 5 line pairs per mm respectively.
- High-contrast resolution is needed for differentiation of breast tissues that have very similar X-ray absorption.
- Wide dynamic range is important, as the image has to include information from both dense regions of glandular tissue, regions of fatty tissue, and all the way out to the skin line.
- Low noise facilitates accurate diagnosis. Noise reduces the contrast resolution.
- Absence of artifacts. Artifacts can for example be caused by electronic noise in the detector, or “ghost images.”
- No dead pixels. Dead pixels can be interpolated from neighboring pixels; this will improve the appearance of the image but not retrieve the lost information.

Low radiation dose

- High quantum efficiency allows the detector to record every incident X-ray photon, maximizing dose efficiency of the system.
- Efficient rejection of scattered radiation. Scatter adds noise and reduces the contrast resolution in the image, which has to be compensated by increasing the exposure.

Available detector technologies

Conceptually, what differentiates the different detector technologies is the number of intermediate conversion steps before a digital signal is formed. This is shown schematically in Figure 1, which indicates the stages at which the signals from individual photons are added (enclosed in green boxes). Each conversion step has the potential to deteriorate image quality. For systems that use techniques other than photon counting, the initial digital signal from the X-rays is converted into an analog signal, and after the exposure one attempts to retrieve the original digital signal through an analog-to-digital converter.

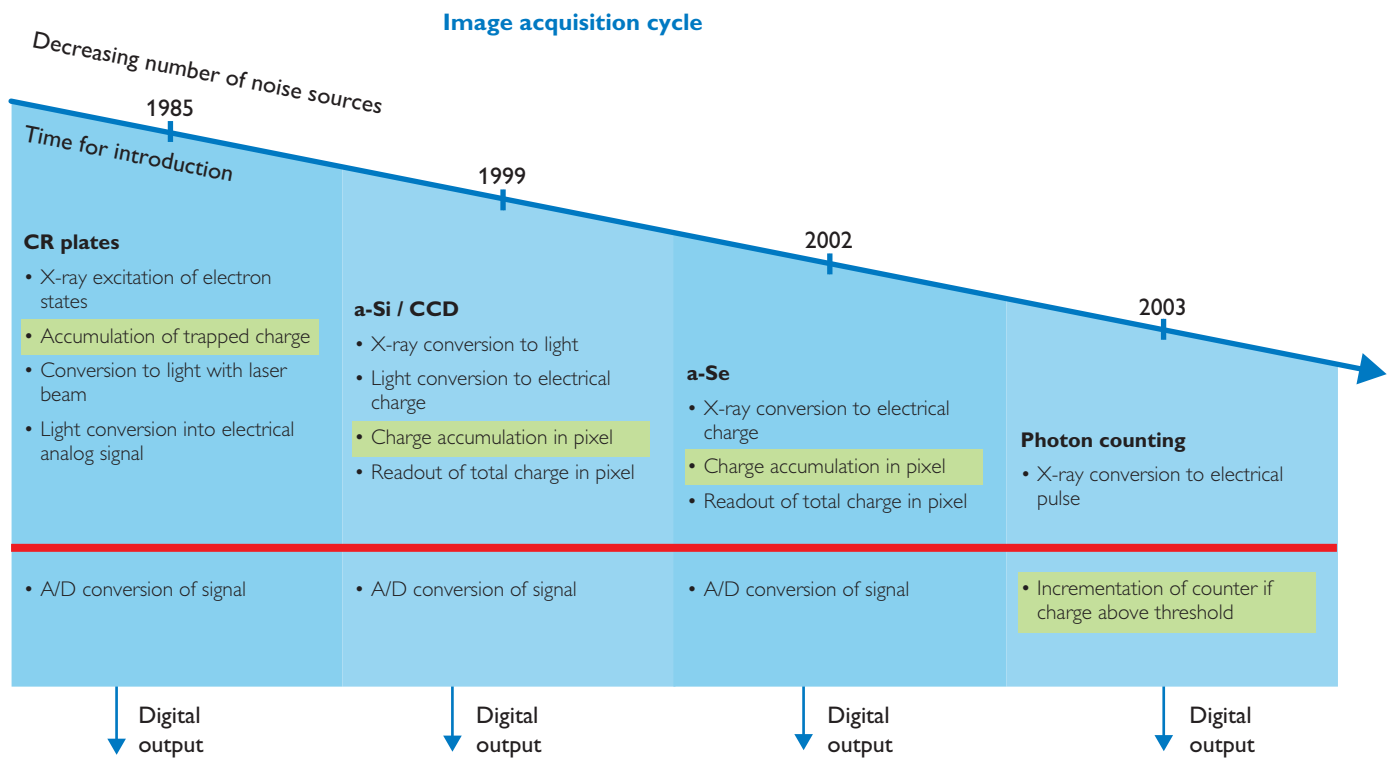


Figure 1 Schematic representation of different digital detector technologies for mammography. Analog steps are above the red line; digital steps are below the red line. The green boxes indicate the stages at which the signals from individual photons are added.

Photon counting technology

X-rays are often perceived as a continuous fluid of energy. In reality this fluid consists of individual particles, so called X-ray photons, which react one-by-one in an X-ray detector. The X-rays are thus digital by nature. The Philips MicroDose detector is built to match this digital nature of X-rays. The Philips MicroDose detector increments the photon counter directly as each individual X-ray is detected. This means that there are no intermediate steps that can add noise from the detection of the photon to the pixel value in the mammogram as indicated in Figure 2. Since there is a risk in every conversion step of adding noise and losing signal, significant effort by several companies has been focused on reducing the number of conversion steps.

The Philips MicroDose detector is a spin-off from high-energy physics research. It is an integrated result from thousands of researchers working from 1960 to the discovery of the top quark in 1996.¹ In high-energy physics, e.g. the search for the top quark, the requirements have been very high for detection efficiency and for extracting maximum information from every detected particle, such as X-ray photons.

Detector Design

The detector system consists of a large number of crystalline silicon strip detectors, with the strips tapered to point back to the X-ray source. Silicon has excellent electronic properties and very good energy resolution. A drawback is the relatively weak attenuation due to the low atomic number ($Z=14$). This is overcome by placing the strip detectors in edge-on geometry with their long axis parallel to the direction of the X-ray beam. In this way, a sufficient absorption length is created, resulting in high quantum efficiency for the full energy spectrum used in mammography.

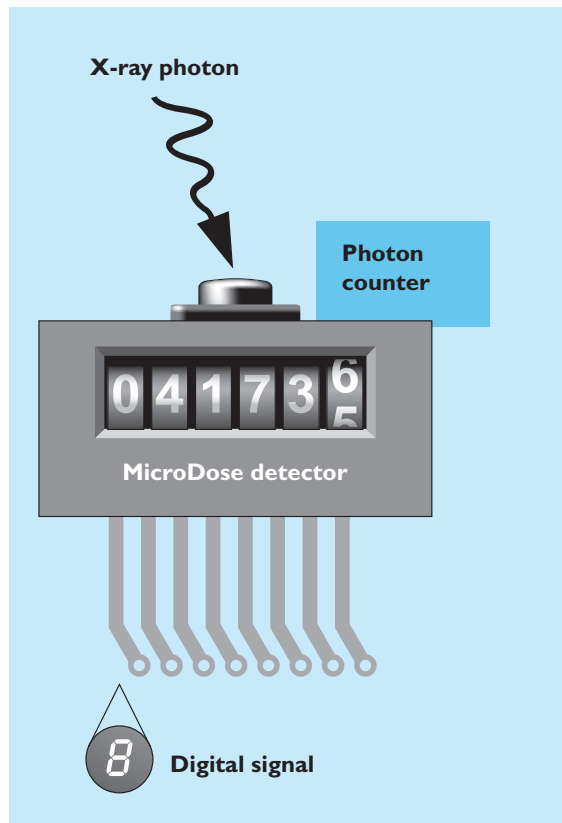


Figure 2 Philips MicroDose photon counting detector.

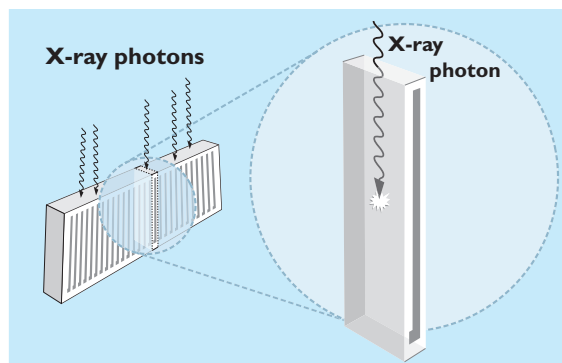


Figure 3 Crystalline Silicon strip detectors with edge-on geometry.

The X-ray interaction in the Si transfers energy to either a photo-electron or Compton electron that subsequently excites electron hole pairs. Since only 3.6 eV is required to create an electron hole pair, several thousand pairs are created for one incident X-ray photon and there is no need for further amplification. A bias voltage is applied across the Si-detector, which generates a bias field. This electrical field causes the charge cloud to be swept towards electrodes with positive voltage and the charge cloud of holes to be swept to electrodes on the opposite side. This drift induces a short electrical signal on the electrodes. The signal is transferred from the aluminum strip to the pulse-shaping electronics through wire-bonds.

Due to the large market for microelectronics that has emerged during the last 50 years, crystalline silicon is by far the purest detector material available. The existing production facilities are huge and standard batch processing of excellent quality and high yield are routine in a reliable production process that has been optimized over the last 4-5 decades.

Readout Electronics

The custom-designed ASIC (Application Specific Integrated Circuit) contains pre-amplifier, shaper, comparator and a digital counter. All channels in the ASIC work in parallel. The shaper creates a signal around a few hundred nanoseconds in duration. The comparator discriminates pulses above a certain threshold level from the detector noise and counts these pulses. The counter has 15 bits to enable wide dynamic range. Each ASIC read-out channel can handle count rates up to 2 MHz, allowing acquisition of images with sufficient statistics in a few seconds. The number of counted photons in each channel is accumulated during a single sampling interval before being sent to the workstation for image reconstruction and display.

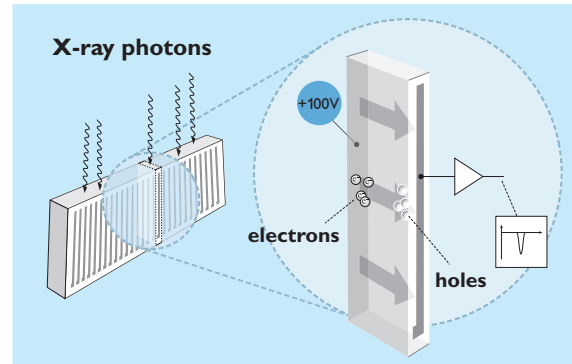


Figure 4 X-ray generates a cloud of electron-hole pairs that create a short electrical signal.

Noise rejection

The pulses created in the Philips MicroDose detector by the incident photons are easily detected, since the pulse height (5000 electrons) is an order of magnitude higher than the electronic noise in the detector (200 electrons rms). Thus the detector counts the number of peak pulses and nothing else. With no conversion steps from X-ray photons to a digital signal; Swank noise, electronic noise and noise from the quantification of the electrical signal from the detector can be removed almost entirely.

If all external noise sources are eliminated, the system is said to work at the quantum limit. This is the ultimate limit for information provided by the laws of physics. Ideally the X-ray photons in all pixels should be distributed according to the mathematical Poisson distribution. These fluctuations are sometimes referred to as quantum mottle or (wrongly) as quantum noise.

For film-screen systems and scintillator-based digital systems, Swank noise can be a serious cause of image degradation. This noise is introduced by the statistical variation of released light after the conversion of the photon.² Hence, two photons of the same energy can have different responses in the detector, which translates to an overall uncertainty in the integrated signal. This effect is to some extent present in integrating detectors such as amorphous silicon detectors.

The absence of Swank noise and the significant removal of electronic noise is a major advantage of the photon counting technology.

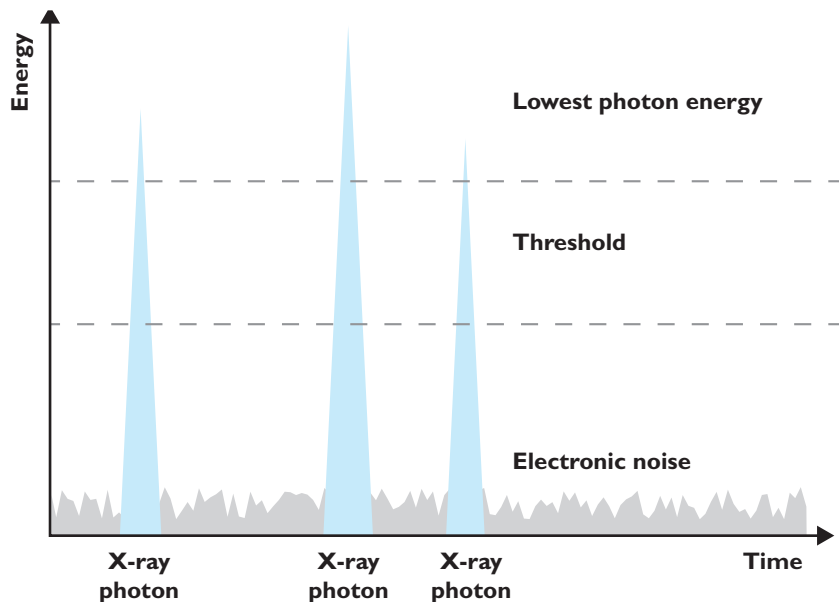


Figure 5 The threshold is set to an optimum level for the energy of the X-ray photons used in mammography, which results in a quantum efficiency of more than 90%.

Weighting of X-ray photons with different energies

To date, all mammography systems available on the market, except for the Philips MicroDose system, have been integrating systems, i.e. the signal is formed by a sum of liberated electric charge or light from X-ray photon interactions. The amount of charge, or light, is proportional to the photon energy. Therefore, with an integrating technique a higher energy photon will gain a higher statistical weight compared to less energetic photons. However, the attenuation in the breast, as well as the contrast, is strongly dependent on energy. It is the low-energy photons that carry more contrast information and should be given a higher statistical weight. This subject was first discussed two decades ago³ but has not generated much attention. The contrast is proportional to the linear attenuation coefficient of the object being imaged.

The subject contrast is roughly proportional to E^{-3} (photon energy), practically independent of tissue type and the size of the abnormality being detected. For integrating systems the signal is proportional to energy, and hence the statistical weight function is incorrect by four powers.

A photon counting system with the same weight for all photons is an improvement over current integration, but does not provide the level of contrast sensitivity afforded by optimal energy weighting. The numerical example to the right is exaggerated to demonstrate the principle. Detailed calculations show that a photon counting system improves the dose efficiency by more than 10%.⁴

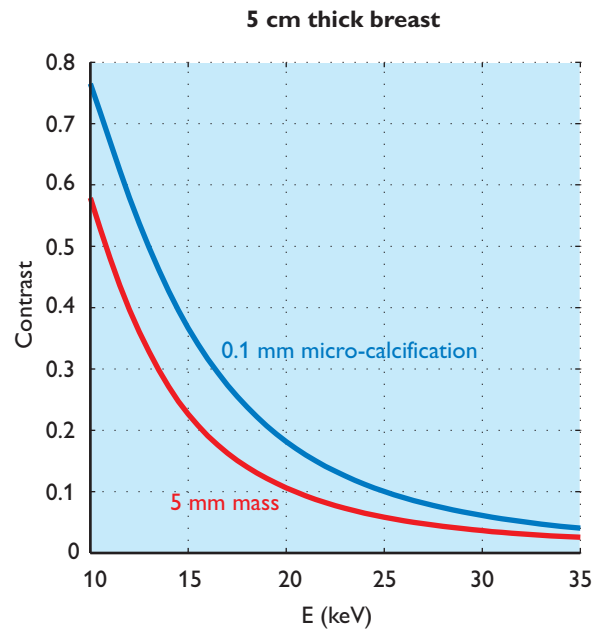


Figure 6 Low energy photons carry more contrast information.

Anti-scatter grids

The conventional way of reducing the scatter-to-primary (S/P) ratio is to insert a Bucky grid between the breast and the detector. The anti-scatter grids are comprised of high aspect-ratio lamellae that absorb the rays that have been scattered at angles beyond acceptance. Depending on the geometry, discrimination between scattered photons and primary photons is more or less efficient. However, for mammography energies, the grids will also remove a significant fraction of the direct photons and are mechanically very difficult to manufacture.

The Bucky factor expresses the increase in patient dosage that is needed to maintain the exposure to the detector when a grid is inserted. It is typically 2-3, and contrast improvements exceeding 80% can be achieved.⁵ The grid usually moves during the exposure to prevent it from appearing in the image, but there are also high strip density, stationary grids. Grids are usually linear, unable to reject photons scattered along the absorbing strips. In recent years, high transmission cellular grids with air interspaces have exhibited improved performance and become more widespread. However, studies show that the use of a grid in conventional mammography provides far from ideal scatter rejection. To quantify, we introduce the scatter DQE as the ratio of squared SNR's between actual scatter rejection and perfect scatter rejection. This can be calculated through:

$$SRE = \frac{T_p}{1 + S/P}$$

where T_p is the transmission of primary photons.⁶ Using values from Ref.,⁷ we find that SRE is in the range 0.5–0.6 for a standard grid.

Scattered radiation

Radiation scattered in the breast is a common source of image degradation in mammography. Scattered radiation will reduce the contrast and increase the noise, which must be compensated by increasing the dose to the patient. This effect is particularly pronounced for thick breasts. It is quantified by the (S/P), which is ideally zero but can be of the order of unity in practice.

Scanning technology and scatter rejection

Achieving efficient scatter rejection is much easier for a scanning system in which a small detector is moved across the object to acquire the full-field image. The main principle is, the smaller the detector used, the lower the risk that a scattered photon will interact. That means no grid is needed in a scanning system and thus a major mechanical simplification is attained since the mechanical construction to achieve a linear translation of a slit is much simpler compared to the complicated mechanics of a Bucky.

Systems are divided into slot and slit scanning systems. For the former, a 2D detector is used, while the latter has a linear detector covering only one pixel row. A multi-slit system has several of these linear slits to create better photon economy, i.e. using a larger portion of the output from the X-ray tube. In this way, the image acquisition time and/or tube load can be reduced.

The load for the X-ray tube will be higher for a scanning system and this is compensated for by increasing the diameter of the anode disc from 3 inches (standard in mammography) to 4 inches, which is a minor modification of the relatively simple X-ray tubes used in mammography.

Philips MicroDose is of the multi-slit type. Photons that are not initially directed towards the detector are removed from the beam by a pre-collimator. These photons are useless and would otherwise only contribute to patient dose. The post-collimator absorbs the photons scattered in the breast and not directed towards the detector. During the continuous scan motion, the detector is read every 2 ms. Besides the cost advantages of using a smaller detector, the main advantage of this technique is the efficient rejection of scattered radiation. The S/P-ratio is only about 3% without any loss of primary radiation.⁸

This inherent scatter rejection is another reason why Philips MicroDose can reduce the radiation dose significantly compared to that used on conventional systems.

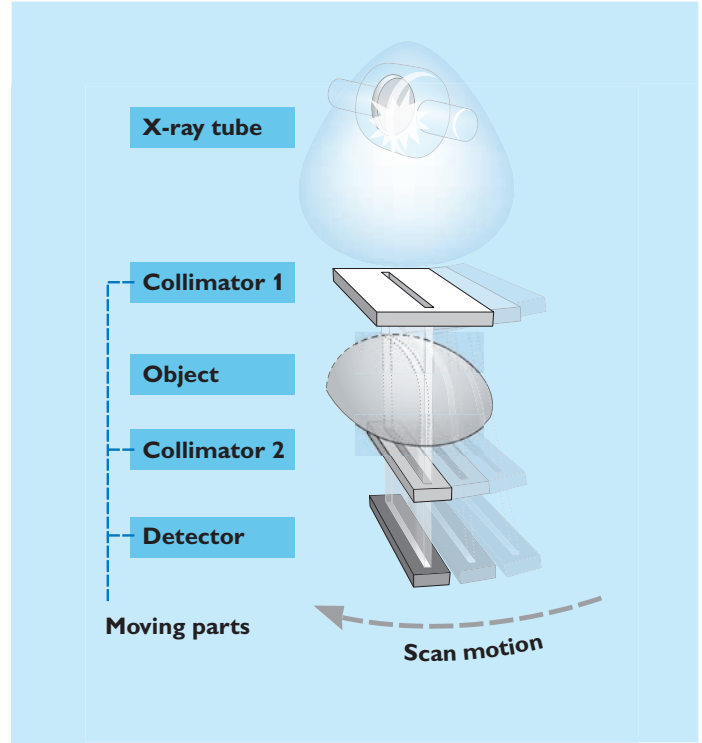


Figure 7 Multi-slit scanning technology used in Philips MicroDose.

System DQE

Scatter DQE

Scatter DQE shows how well the system rejects scattered radiation from the breast. Detection of scattered X-rays decreases the contrast and degrades image quality. Conventional Bucky grids are far from perfect and have a scatter DQE of roughly 50%. The scanning multi-slit geometry of Philips MicroDose provides an excellent scatter rejection and a scatter DQE of 96%.⁸

Energy weighting

Energy weighting describes how the information in the X-ray beam is captured by the detector. The low energy photons contain more information than the high energy ones, and should thus be weighted higher by the detector. Energy-integrating systems do the opposite, while photon counting detectors put equal weight on each photon. The energy weighting factor shows how much efficiency is lost compared to a photon counting system.⁴

Traditionally, X-ray detectors have been compared through the detective quantum efficiency (DQE). With the advent of new technologies, such as photon counting and multi-slit-scanning, detector DQE is an inadequate measure of the complete imaging system's performance. A better figure-of-merit is the system DQE, in which several additional properties have been taken into account. The system DQE is composed of three factors:

- Detector DQE
- Scatter DQE
- Energy weighting

Detector DQE

Detector DQE is the conventional measure of how a regular pattern is imaged by the system and especially shows how this is degraded by noise in the system. Degradation by scattered radiation is not included here. Neither does this measure show how contrasts in actual breast tissue are imaged.⁹

Summary of characteristics for Philips MicroDose

- Matches the digital nature of X-rays; each X-ray is processed individually and external noise is almost entirely eliminated High Quantum Efficiency
- Almost perfect scatter rejection
- 50 μm pixel size
- Field of view 24 x 26 cm
- Scan time 3-15 seconds
- Tungsten anode, 0.5 mm Al filter
- All radiation hits active detector area, no radiation “spill”
- Maximum count rate 2 MHz per pixel and no “ghost images”
- No dead pixels

References

- 1 Abe, F. et al., 1994. Evidence for Top Quark Production in pp Collisions at $\sqrt{s} = 1.8$ TeV. *Fermilab-PUB-94/116-E*. May 16, pp.1-19
- 2 Swank, R., 1973. Absorption and noise in X-ray phosphors. *Journal of Applied Physics*, 44, pp.4199–203.
- 3 Tapiovaara, M., Wagner, R., 1985. SNR and DQE analysis of broad spectrum X-ray imaging. *Physics in Medicine and Biology*, 30, pp.519–29.
- 4 Cahn, R.N., Cederström, B., Danielsson, M., Hall, A., Lundqvist, M., Nygren, D., 1999. Detective Quantum Efficiency dependence on X-ray energy weighting in Mammography. *Medical Physics*, 26(12), pp.2680-3.
- 5 Rezentes, P., de Almeida, A., Barnes, G., 1999. Mammography grid performance. *Radiology*, 210(1), pp. 227–32.
- 6 Wagner, R., Barnes, G.T., Askins, B.S., 1980. Effect of reduced scatter on radiographic information content and patient exposure: A quantitative demonstration. *Medical Physics*, 7(1), pp.13-8.
- 7 Dance, D.R., Persliden, J., Carlson, G.A., 1992. Calculation of dose and contrast for two mammography grids. *Physics in Medicine and Biology*, 37(1), pp.235-48.
- 8 Åslund, M., Cederström, B., Lundqvist, M., Danielsson, M., 2006. Scatter rejection in multi-slit digital Mammography. *Medical Physics*, 33(4), pp.933-40.
- 9 Dobbins III, J.T., 2000. Image Quality Metrics for Digital Systems. In: J. Beutel, H.L. Kundel, R.L. van Metter, ed. 2000. *Handbook of Medical Imaging, Volume 1. Physics and Psychophysics*. Bellingham WA, USA: SPIE Press. pp.161-219



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