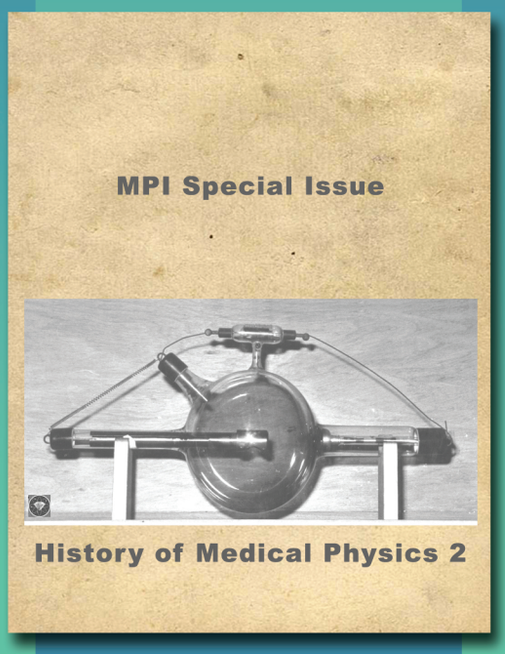
MEDICAL PHYSICS
International



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MPI

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The Journal of the International Organization for Medical Physics

Aims and Coverage:
Medical Physics International (MPI) is the official IOMP journal. The journal provides a new platform for medical physicists to share their experience, ideas and new information generated from their work of scientific, educational and professional nature. The e- journal is available free of charge to IOMP members.

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EDITORIAL

Slavik Tabakov and Perry Sprawls

MPI Co-Editors

Welcome to the second Special Issue of Medical Physics International which is devoted to articles developed in the IOMP History Project. The topics from this issue will be presented in the History Symposium at the 2019 AAPM Annual Meeting in San Antonio, Texas (July 2019). This results from the collaboration of the AAPM History Committee with the IOMP History Project. Under the direction of the AAPM History Symposium Working Group future Symposia will include presentations from both the History project and other relevant presentations showing the developments in medical physics and the place of our profession in contemporary healthcare

The first Special Issue of the Journal Medical Physics International (MPI SI 1, May 2018) has over 5000 downloads. It can be downloaded free at: http://www.mpijournal.org/pdf/2018-SI-01/MPI-2018-SI-01.pdf.

The topics in the First Edition are:

- X-ray Tubes Development - IOMP History of Medical Physics (R. Behling) p.8
- Film-Screen Radiography Receptor Development – A Historical Perspective (P Sprawls) p.56
- History of Medical Physics e-Learning Introduction and First Steps - IOMP History of Medical Physics (S Tabakov) p.82

This second Special Edition here adds more chapters to the History project.

After the official announcement of the History project in the MPI Journal (May 2017) - available at www.mpijournal.org/pdf/2017-01/MPI-2017-01-p068.pdf - the number of chapters in the project is growing. This could result to more frequent Special Issues, which will later be combined in volumes related to the specific sub-parts of medical physics and the development of the profession.

As stated before, the project chapters will be open ended in order to further expand in future with more methods and equipment, showing the contribution of thousands of medical physicists, the medical technology industry and various organisations to the global healthcare progress. To facilitate the project progress we shall regularly present information about the History project in the MPI Journal and other publications. We welcome the contribution of colleagues from all societies, organisations and companies to join the History project in its various volumes.

We look forward to your contributions to the Project

Fluoroscopic Technology from 1895 to 2019
Drivers: Physics and Physiology

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Abstract: X-rays were first observed when they caused a phosphor to fluoresce in Roentgen's laboratory. In the early years, X-ray tubes and their associated power supplies were unstable and often dangerous to use due to X-rays, exposed high-voltage wiring, and ozone. The hazards were well known. Even so many early fluoroscopists lost their hands or lives. Nevertheless, by 1900, fluoroscopy became a routine medical tool. Medical radiography evolved as well but was slowed by long exposure times and the difficulties of early 20th century photographic processing. By the 1940s, the technology had reached the level at which the equipment required little attention, permitting the operators to focus almost exclusively on the medical tasks. Several key papers from this time period discuss noise effects, the effects of dark adaption on visual perception, the need for brighter images without excessive radiation, and begin to introduce cognitive factors into image diagnosis. The X-ray image intensifier (II) introduced into practice in the 1950s was a major step toward addressing these problems. The II also produced enough light to permit photographic and cine recordings of its output. The II itself achieved a factor of two improvement in quantum efficiency when the ZnCdS input screen was replaced by CsI in the mid-1970s. Parallel developments in video technology enabled the deployment of video viewing as a replacement for direct optical viewing. Photographic image subtraction was a key aspect of angiography. By 1980, analog and digital subtraction of angiographic images was a key factor in the growth of interventional radiology. Improvements in X-ray tube technology yielded greater outputs which facilitated spectral shaping of the X-ray beams to better match the absorption properties of Iodine. This step simultaneously reduced patient irradiation and improved the visibility of contrast filled vessels. The image intensifier itself began to be replaced by solid-state detectors (FP) around 2000. The most common variant used, and still uses a CsI input layer that is similar to that found in IIs. Thus, the dosimetric characteristics these FPs is similar to IIs. By 2010, the imaging hardware had become relatively stable. Improved real-time image processing has enabled better coupling of the information in the image to the observer's eye. This has yielded further improvements in image utility while simultaneously reducing patient irradiation. The next decade is likely to see further advances in this direction.

Keywords— Fluoroscopy, interventional radiology, patient dose, image quality, image processing

CONTENT

I. INTRODUCTION

The evolution of fluoroscopic technology has driven by clinical requirements, technological possibilities, and the operational necessities of real-time imaging. Fluoroscopic procedures are used to monitor the motion of clinical objects using sequences of transitory images. Supplementary technologies are used to acquire and preserve a fraction of these images for later analysis. Most fluoroscopic procedures use real-time imaging while manipulating the patient, objects within the patient, and the imaging geometry needed to optimally view these objects. This process exposes the patient to radiation risks. Fluoroscopy frequently requires the presence of operators close to the patient and results in their exposure to radiation risks as well.

Fluoroscopy always requires that operators play a continuously active role while a procedure is in progress. The operator's physiological requirements, knowledge of the patient's condition, and immediate visual feedback provide key real-time operational inputs to the fluoroscope. This is fundamentally different than radiography, where images are usually acquired using a predefined protocol and there is seldom any interaction between physician and patient during clinical utilization of the images.

Key non-hardware concepts needed to understand the fluoroscopic environment include the physiology of human vision, the presence of a noise-defining quantum sink in the imaging chain, and the use of image processing to improve the conspicuity of clinically important structures. Many of the technological jumps that occurred in the past century are direct responses to requirements imposed by these concepts.

II. EARLY DAYS (1895 – 1920)

Roentgen's sequence of three original papers [1-3] and other early sources outline both similarities and difference between fluoroscopy and acquisition. For example: Roentgen's first observation was the fluorescence of Barium platinocyanide on a piece of cardboard when his Crookes tube was powered. It was reported that he saw a transient image of his own hand on the screen while manipulating an object with the beam on. The radiograph of his wife's hand was probably produced to document this observation.

Crookes tubes, their associated high-voltage sources, and other relevant items, were available in many physics laboratories around the world when Rontgen's discovery was announced in December of 1895. Experiments, including fluoroscopy, were reported around the world within weeks. Given the number of reports in the following few months, everything needed to produce fluoroscopic images must have been immediately at hand in many physics laboratories. Medical X-ray imaging ensued in many of these venues within weeks of the announcement. An anonymous research review was published in Nature in June 1896 [4]. It starts with: "The novelty of Prof. Rontgen's skeletal photographs has almost worn off, and the field of research opened up by his observations is now mainly occupied by scientific workers …" Nevertheless, there was continued physicist interest in X-ray imaging. Figures 1 and 2 document two anonymous nineteenth century setups with an induction-coil as the high-voltage source and a cryptoscope as the fluoroscopic detector. The tube in Figure 1 appears to be a basic Crook's design. The tubes in Figure 2 may already be contain an anode and a concave cathode. These features serve to increase X-Ray output and sharpen the image.

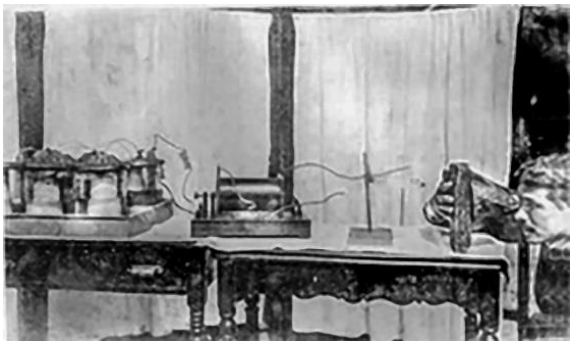


Fig 1: c 1896 Physicists' fluoroscope.

The monocular cryptoscope is the only item that may not have been immediately available in a typical physics lab following Roentgen's announcement in 1895. The investigator is using his own hand as a test-object. There were several published reports of hand injuries before the end of 1896. Source - ANON

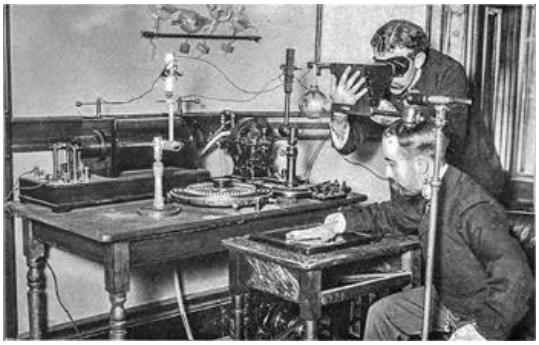


Fig 2: 1899 Radiography and Fluoroscopy.

This ubiquitous photo of a seated individual's hand is being radiographed while the standing individual is simultaneously using a cryptoscope to fluoroscope his own hand. This image could be either a posed picture or the fluoroscopist is using his hand as a QA tool while he adjusts the apparatus. . Note the supply of additional X-ray tubes on the far wall. Source – ANON

As shown in Figures 1 and 2, many early radiation workers used fluoroscopy with human hands as test objects with catastrophic consequences. Among other 1896 reports is an extended first-person description of major tissue reactions on the author's hands published in October [5].

X-rays outputs from gas-discharge tubes are unstable. Roentgen described a tool for describing imaging characteristics of the beam in his third paper [3]. It is simply a sheet of platinum with an array of holes. Each hole is covered with a different thickness of aluminum. The images of the tool provide the thickness of aluminum needed to match the transmission through the surrounding platinum. This tool is a variant of the Bunsen photometer used for measuring light intensities in the pre-electronic era. The underlying physics of atomic number dependent differential attenuation is the basis of many 20th and 21st century test tools. Many other fluoroscopic test objects, and reference objects for characterizing patient images were described in the literature before 1900.

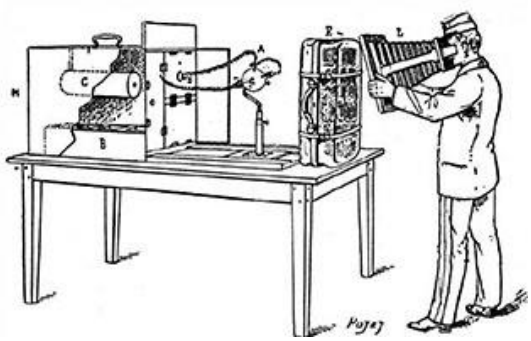
Direct observation of the fluoroscopic screen involves directly converting a fraction of the energy carried by an X-ray beam into visible light (fluorescent-yield). Image visualization requires enough light to activate either the human visual system or a photographic plate. In the weeks after receiving the first news about Roentgen's discovery, Edison screened a vast number of compounds for their fluorescent-yield and reported calcium tungstate (CaW) as the optimum in 1896 [6]. Ba[Pt(CN)4] produces green fluorescent light, and CaW produces blue light. The dark-adapted human eye is more sensitive to green than blue. CaW became the phosphor of choice for radiographic intensifying screens. Fluoroscopic screens eventually used green-emitting zinc-cadmium-sulphide (ZnCdS) as the phosphor of choice. ZnCdS was also used as the input screen for the first few generations of image intensifiers, until it was replaced by CsI in the 1980s.

The same report [6] goes on to say "The importance of this [fluoroscopic] apparatus to the surgeon cannot be over-estimated. It will give him an instant diagnosis of his case. The photographic method involves long exposure, in itself an evil, followed by the slow development and drying of the plate, and, worst of all, the uncertainty of getting any result whatever. The fluoroscope tells the story at once." Edison was a very active investigator of X-rays from within weeks of Roentgen's announcement until his assistant Clarence Dally became the first fatality from radiation in 1904 [7]. Because of this, Edison totally stopped all work with X-rays with the following comment: "Don't talk to me about X-rays, I am afraid of them."

In the nineteenth century most X-ray imaging was fluoroscopy using a hooded enclosure or a hand-held open fluoroscopic screen as the image receptor. The hooded system is called a cryptoscope in this paper to distinguish this technology from other types of fluoroscopic systems.

Figures 3 and 4 1897 show fluoroscopy in use for customs inspections. These are from the same Scientific American article [8]. The photograph in Figure 3, indicates that the working environment was illuminated.

Fig 3: 1897 baggage inspection.



Source – Scientific American – Permission Pending

Figure 4 is a drawing of an open fluoroscopic screen used to search a woman for contraband in a darkened room. The X-ray components (box, tube, tube-stand) are identical to those in Figure 3. Photography in the dark was simply not possible in the 19th century. This image has also taken on its own life over the years; one version is captioned 'x-ray parlor games'

Fig 4: 1897 open fluoroscopic screen

Customs inspection detecting contraband. The woman on the right appears to be a badged customs officer. This image was included in the article cited in Figure 3. It has been reproduced many times with captions such as "X-ray parlor games".

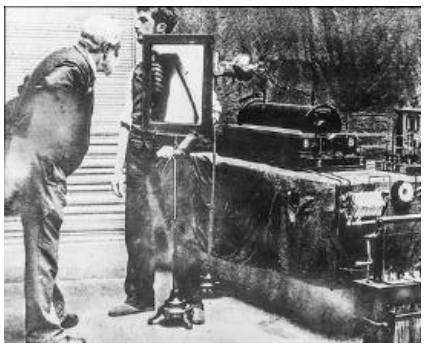
Source – Scientific American – Permission Pending

Figure 5 contains a photograph of an early fluoroscopic examination. The source of this image is unknown but must be prior that of the almost identical French adverting postcard (Welcome Trust) The X-ray system seen in the images has essentially the same components as the system shown in Figure 1. The 19th century equivalent of photoshop must have been used to produce the 'original' photo on the left. The fluoroscopic image of the patient is far brighter than possible with direct fluoroscopy and two left elbows are seen (one on the screen) and one outside the boundary of the screen).

Fig 5: Pre 1900 Chest Fluoroscopy

a) Photograph with 19th century 'photoshop'. ANON

b) c 1898 advertising postcard. Source -Welcome Trust – permission pending

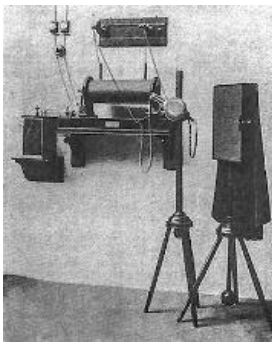


A

B

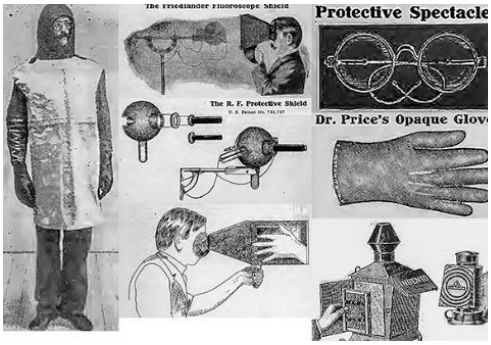
Fig 6: 1896 Complete X-ray system.

The unshielded X-ray tube is just right of center There is an attached spark-gap voltmeter mounted on the wall. The image receptor consists of a large fluoroscopic screen. A black cloth hood like the hood used photographic cameras of the era isolates the operator from room light. Source – Siemens Healthineers.

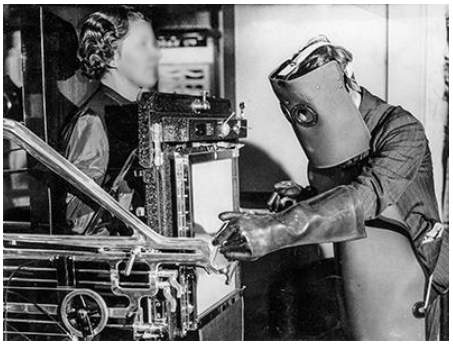


Complete X-ray sets were soon offered by major electrotechnical companies such as the Siemens set shown in (Fig 6). There were many other firms providing X-ray equipment and supplies c 1900. The surviving 1905 third edition catalog from Friedlander is fascinating. Beyond tubes, tube-stands, and generators this catalog provided instructions for battery-fluid film-processing chemistry, and assortment of radiation protection materials. Examples of radiation PPE are shown in Figure 7. The need for these devices was driven by reports of hand damage as early as 1896[9] and the death of Clarence Daily (Edison's assistant) in 1904[7].

Fig 7: Radiation Protection Devices



a) Selected offerings from 1905 Friedlander catalog No. 3. Source – Public Domain



b) c 1940. What the well-dressed fluoroscopist might wear Source - ANON

Electrotherapy equipment was commercially manufactured and distributed in the late 19th and early 20th centuries. These devices used high-voltage sources that could produce X-rays in an increasing variety of gas-discharge tubes. By the turn of the century, McIntosh Battery & Optical Company a manufacturer of electrotherapy equipment, was able to supply commercial X-ray sets by adding a tube and cryptoscope to their existing product line. Fig 8 reproduces a 1915 advertisement from this company. Despite the image of an irradiated hand in the ad, this manufacturer was aware of the dangers of using the operator's hand as a QA tool. Commercial fluoroscopic QA tools have been around for more than a century. McIntosh offered a multi-target penetrometer with each target characterized as 'inches of flesh'. The long handle allowed the tool to be used without needing a hand in the beam. A similar penetrometer (without the handle) was offered in the 1905 Friedlander catalog.

Fig 8: 1915 Fluoroscopic set and an accessory QA tool

See text for the description.

Source - ORAU Health Physics Historical Museum – permission pending

Early in the gas-tube era, it was stated that radiography added no value to the fluoroscopic examination. There was likely to be partially true because that required long radiographic exposure times (typically tens of minutes) often resulted in motion blur. Photographic processing of radiographs was also in its infancy. Unfortunately, many early fluoroscopists were not content with brief observations; this resulted in many instances of radiation injury of both patients and operators. A decade or so later, better X-ray sources and improved radiographic receptors changed to advise that fluoroscopy should be only used for procedures requiring the observation of motion. Figure 9 reproduces two 1910 cryptoscopic head examinations[10]. Skull fluoroscopy was rare in this era. Figure 10, from the same textbook edition illustrates documentation using notes about or tracings of the fluoroscopic images. This textbook specifically advises that "radiography should be used for documentation instead of fluoroscopy whenever possible."

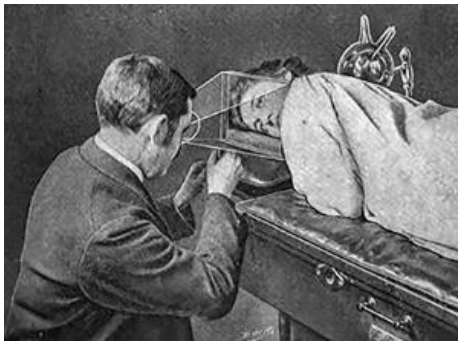


Fig 9: 1910 Textbook illustrations
The glass bowl supporting the gas X-ray tube is an electrical insulator. The X-ray tube in the left image is contained in a glass bowl that appears to have a partial lead shield similar to the shield offered in the Friedlander catalog (Fig 07). Note the exposed high-voltage wiring. The right image shows an unshielded X-ray tube. Source - Tousey 1910

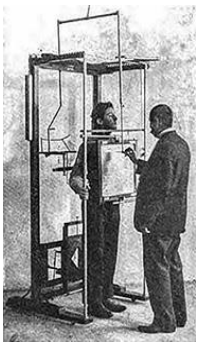


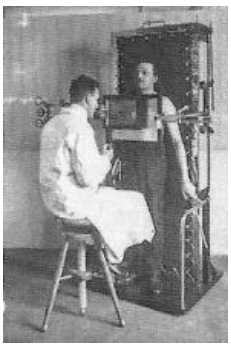
Fig 10: 1910 Image documentation Captions in original advises to use radiography instead of notes or sketches based on fluoroscopy for documentation. Source - Tousey 1910

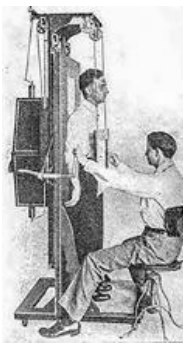
The need to have one or both hands free to manipulate the patient, adjust the X-ray controls, and document the imaging findings resulted in mechanical fluoroscopic screen holders attached to some form of patient support. The evolution of patient support mechanisms is a key part of fluoroscopy history from 1900-1950. The critical elements were anatomically positioning the patient for the procedure and giving the physician appropriate access to the patient during the procedure. Figure 11 illustrates two such systems. From the dates of the pictures, the X-ray sources were gas tubes. Along with radiation, exposed high voltage wires and ozone production were real dangers. These goals had to be met for procedures that were done in complete darkness. As time progressed systems incorporated improvements in both electrical and radiation protection for patients and staff.

Fig 11: Vertical Fluoroscopes

a) (1907) "the Kinescope allowed fluoroscopic examination and X-ray photography of patients standing, sitting, or lying with only one device." Source - Siemens Healthineers

b) Lead screen cover over the X-ray tube indicated in caption. Source – Tousey 1910





As shown in Figure 12, fluoroscopic guided surgery using a gas-tube and a head-mounted cryptoscope was in use for fluoroscopically guided surgery in WW-I. This is an early example of a fluoroscopically guided interventional procedure. Cryptoscopes remained in military and civilian use through the 1940s for both medical and non-medical purposes.

Fig 12: Military cryptoscopes

circa 1916

1944 US Navy Art Collection

1944 Europe

III. DIRECT FLUOROSCOPIC SCREENS (1905-1960)

The gas-tube era ended in the later part of the 1010's when hot-filament X-ray tube and transformer driven power supplies replaced gas tubes and induction-coil/static-generator high-voltage sources. The operational simplicity of these newer components helped transition daily use of radiography and fluoroscopy from a focus on equipment technology to a focus on radiation and imaging.

The transitions from gas-tubes and hand-held screens occurred in stages. Figure 13 is from Acme's 1926 advertisement for a horizontal table. The fluoroscopic screen is above the table. The poles on the left are tie-points for the high-voltage wiring. The radiographic tube holder (minus the tube itself-probably it was a hot-filament tube) is over the table with its fixed collimator. The tube is mounted in a lead-glass bowl which served both as a radiation barrier and an electrical insulator. The fluoroscopic tube is encased in a light-tight enclosure under the table. This enclosure may or may-not have had lead shielding. There is exposed high voltage wiring over and under the table. Nevertheless, this apparatus was used for fluoroscopy in a completely dark room.

A comparable 1940 Picker table is shown in Figure 14. Protection against electrical hazards with the introduction of flexible insulated high-voltage cables. The enclosed table contributed to reducing the amount of scatter reaching the operator. Equipment mounted specific radiation protection elements are missing.

Fig13: 1926 horizontal table

This Acme system includes an over-table radiographic tube (with collimator) and an under-table fluoroscopic tube. The radiographic tube itself and its wiring are not shown in this image. The two poles on the left-hand side are insulators. In use, exposed high-voltage wires ran from these points to the X-ray tubes. The enclosure for the under-table fluoroscopic tube may have been provided to minimize stray light during fluoroscopy.

Source - ACME advertisement in RADIOLOGY

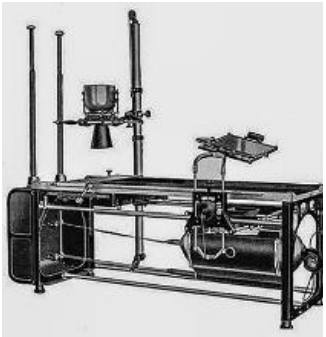
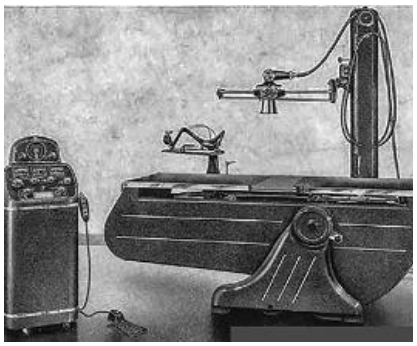


Fig 14: 1940 R/F table

Insulated high-voltage cables replace open wiring. The X-ray tubes used for such systems were typically shielded by lead contained within a grounded metal casing.

Source - PICKER advertisement in RADIOLOGY



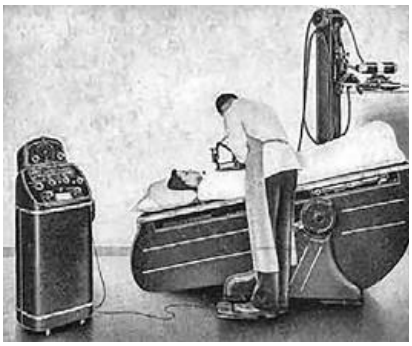
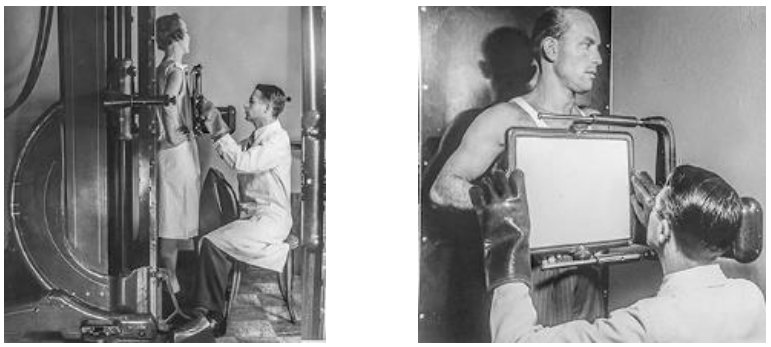


Figure 15 illustrates chest fluoroscopy in 1947. The original source of these images has been lost. Both appear to show the same fluoroscope. By this time, the fluoroscopic-screen may have had a lead-glass covering providing radiation protection to the operator. No collimation controls are immediately evident in either photo. If collimation was not available, the beam size would have been large enough to fully illuminate the screen in any position. The fluoroscopic screen is linked to the X-Ray tube assembly only in the vertical dimension. Thus, the vertical size of the beam need not have exceeded that of the screen. However, on the right-hand image, the screen is free to slide in the horizontal direction. This implies that the beam's horizontal size was substantially larger than that of the screen. This is undesirable. Eventually regulations required both collimation and a visible unirradiated margin on all four sides of the image. The heavy lead glove worn on the left-hand was used to palpate the patient under fluoroscopic control. It is also noted that the operator is not wearing any form of body radiation protection. There appears to be a lead apron draped over the chair back in the left-hand image. A portion of the scatter produced in the patient's chest could pass through the unshielded are between the bottom of the screen and the chair and thus irradiate the operator. Later standards specified a fully protected zone for the operator to manage irradiation.

Fig 15: 1947 Chest Fluoroscopy

See text for description

Source - ORAU Health Physics Historical Museum – permission pending



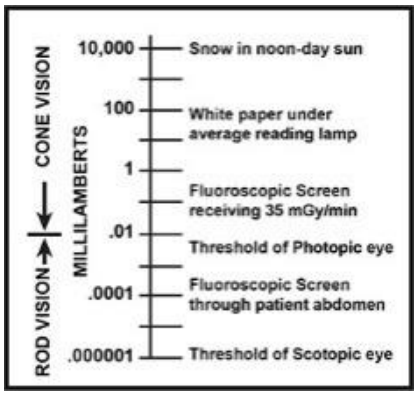
IV. THE OBSERVER'S EYE

The image on a fluoroscopic screen is only of value when it is transmitted to the observer's brain via the eyes. Aspects of vision intimately related to fluoroscopy are discussed in this section. The 10^{10} working light-intensity range of the human eye is needed to accommodate mid-day light levels in a cloudless desert and a subsequent moonless night(Fig 16). The anatomical basis providing this range includes two types of photoreceptors in the retina. The cone cells are activated by high light-intensity, the rod cells work at low light-intensity. Additional information is provided by standard vision and perception textbooks [11, 12].

Fig 16: Range of light intensities over which the human eye can accommodate

This range includes the effects of dark adaptation

Adapted from Chamberlain – 1942 [13]



The increase in sensitivity of the eye to dim light is called dark adaptation. This process takes tens of minutes as shown in Figure 17. Note the break at the level where the cones have reached their maximum sensitivity, and the extension of sensitivity by several orders of magnitude where rod vision dominates.

Fig 17: Sketch of eye sensitivity as a function of dark-adaptation time.

The retina contains a mixture of visual cells called rods and cones. The cones are concentrated in the fovea, and rods compose the rest of the visual field.

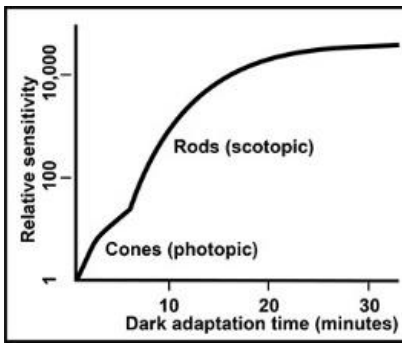


Figure 18 is from around 1955. Radiologists had to dark adapt for tens of minutes in order to optimally use their fluoroscopic screens. The radiologists are wearing red goggles which allows adaptation in typical ambient illumination. Radiographic film images could not be adequately read or interpreted while dark adapting. The goggles were removed after going into the fluoroscopic room and switching off the lights. There are many available photographs showing radiologists using their fluoroscopes while wearing red goggles. This is an impossibility because the screens emitted green light.

Fig 18: c 1955 dark adaptation with goggles.

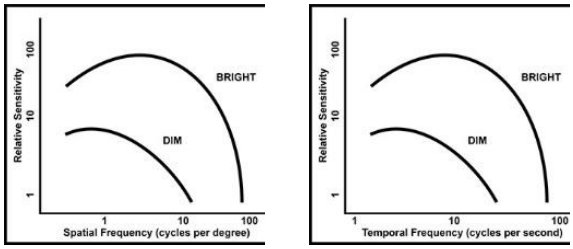
Fluoroscopic screens could only produce a limited luminance at acceptable dose-rates. Operators dark adapted for about 30 minutes in order to fluoroscope at these low light levels. Typically, they wore red goggles and avoided bright lights during the adaptation time.

Source – ANON

Working with dark adapted eyes reduces the amount of radiation needed to form a perceivable image. Fortunately, the X-ray tubes available in the fluoroscopic-screen-era could not produce high dose-rates for long periods of time without sustaining irreversible thermal damage. Observers had to dark adapt in order to see anything of value. However, the costs of increased sensitivity include decreased spatial and temporal resolution of the visual system (Fig 19).

Fig 19: Sketches of spatial and temporal resolution at photopic and scotopic light levels.

Bright illumination levels enable foveal vision with the cones resulting in better spatial and temporal resolution. The present specification for minimum illuminance on diagnostic image monitors is intended to drive foveal vision.



V. RADIOGRAPHY BY PHOTOGRAPHY OF FLUOROSCOPIC SCREENS

The concept of photographing a fluoroscopic screen was proposed before 1900. It was not practical then due to the limited light output of screens and the low sensitivity of early films. With better film and optics, photofluorographic systems, used for mass chest radiography were in common use from the 1930's through the 1960's [14]

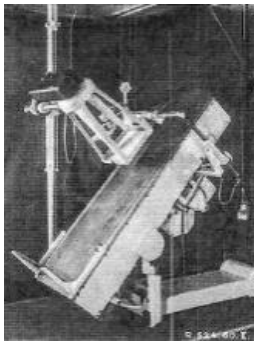
Several of the surviving photographs (i.e. Fig. 10) indicate that sketching on top of the fluoroscopic screen and note-taking were in use. Several tools using direct irradiation of film-screen systems (spot-film-devices, kymography, 'rapid film changers') were used up to the 1980's but are mostly beyond the scope of this paper.

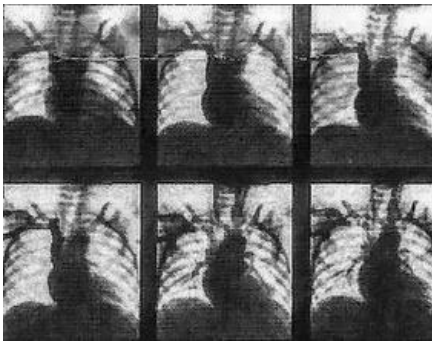
Rapid series imaging with a direct fluorescent screen was possible but not common. A 1950 system is shown in Figure 20. Based on the sample coronary angiogram, this system operated at several frames per second. The X-ray tube loading needed to sustain this framerate, and patient dose resulting from photographing direct screens, were too high to permit many frames, this inhibited general acceptance of this technology.

Fig 20: 1950 rapid sequence angiographic system

Uses a fluorescent screen and a photographic camera. This configuration was seldom used clinically. Along with limited image quality, patient radiation doses were almost certainly much higher than film-screen radiography.

Source - Siemens Healthineers





VI. NOISE LIMITED IMAGING

The perception of key clinical objects in an image is influenced by image noise. In a fluoroscopic image, visual noise can be grouped into two categories: The first is the noise produced by statistical variations in the numbers of X-Ray photons converted into light by the fluoroscopic screen (Quantum Noise). The second category is the combination of all other noise sources in the imaging chain and the observer's eye-brain system (System Noise). Quantum Noise is dependent on the local X-ray dose. System noise is independent of X-ray dose but may be dependent on factors such as illumination level. The amount of noise in an image changes with dose only when Quantum Noise is dominant.

For fluoroscopic-screens, quantum noise is significant if not totally dominant at patient entrance dose-rates of tens of milligray per minute. At any dose-rate. a device that amplifies illuminance of the observer's retina simply provides a brighter image without affecting quantum noise. Quantum noise is increased if the X-ray dose-rate is reduced. Low dose fluoroscopy may be far too (quantum) noisy for clinical use.

Higher dose-rates are required for direct fluoroscopy if the eye is not appropriately dark adapted. Under these circumstances, additional X-ray energy is needed to produce enough light for the observer to see. In broad terms, good screen-fluoroscopy was performed at dose-rates comparable to modern systems. Maximum dose rates were likely to have been physically restricted to a few hundred mGy/min by the construction of the X-ray tubes of the era.

Information flow through a fluoroscopic imaging system has been represented by a photon-flux-diagram since the 1940's [15]. Figure 21 introduces this model. The key concept is that radiographic and fluoroscopic images contain noise deriving from the quantum nature of the X-ray beam and modified at each step in the imaging chain. The ordinate is the number of photons per unit area. Several steps in the process are shown:

1) X-ray photon flux at the patient's skin (entrance exposure rate without scatter).
2) X-ray photon flux at the fluoroscopic screen (normalized to 1.0 for trajectories A and B)
3) Detected X-ray photons (25% of incident)
4) Light conversion (1,000 useful light photons per detected X-ray photon)
5) Useful light photons at the observer's eye (set to 2.5 for trajectories A,C and 0.25 for B)

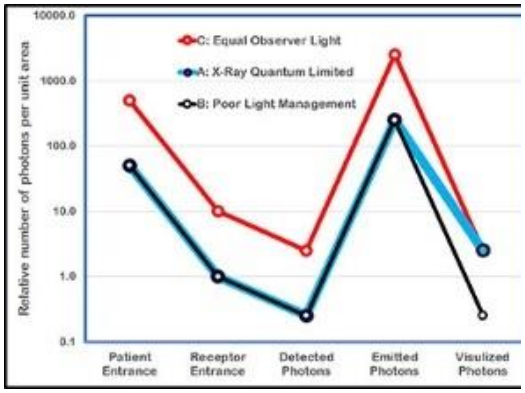
The smallest number of signal carriers in a trajectory is called the quantum sink. In the quantum-limited trajectory (A), the number of visualized photons is an order of magnitude higher than detected X-Ray photons, indicating that X-ray quantum noise will dominate. The number of visualized photons in the poor-light management trajectory (B) is the quantum sink and is of concern. Since it is equal the number of detected photons on the same trajectory, it indicates that visualized noise will be higher than the native quantum noise. The equal light trajectory (C) delivers the same number of visualized photons as trajectory A, but at the cost of 10 X the patient input dose. Much of the technical development of image receptors in the subsequent half-century was focused on maintaining the quantum sink at the detected photon level instead of elsewhere in the imaging hardware.

Fig 21: Fluoro screen photon-flux diagram

A) QUANTUM LIMITED: The photon-flux density at the eye is 10 X the detected level. Entrance Exposure Rate (EER) = 50

B) POOR LIGHT MANAGEMENT: The photon-flux density at the eye is equal to that at detection stage. Visible noise is due to a combination of X-ray quantum noise and 'poor counting statistics' at the observer's eye. EER = 50.

C) EQUAL OBSERVER LIGHT (using the same ratio of emitted to visualized photons noted for B): Patient EER is increased by a factor of ten under these conditions EER = 500

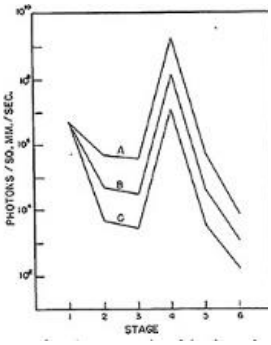


Sturm and Morgan's [15] original diagrams are reproduced in Figure 22. Patient entrance dose is constant for all three trajectories. Patient exit dose decreases for thicker beam projections. These change impact the downstream photon fluence density. The loss of flux inside the eye is an additional stage to those shown in Fig 21. This diagram indicates that the quantum sink is the observer's retina. The contrast-detail curves on the right are related to spatial sensitivity curve in Fig 19 with image noise as an additional driver. These experiments assumed a constant patient entrance exposure rate (left) Note the effect of increased image noise (due to lower output light flux) in trajectory C compared to trajectories A and B. In the right-hand drawing, the solid curves are experimental, and the dotted lines are corresponding theoretical predictions.

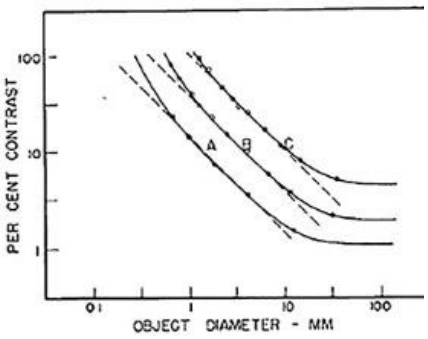
Fig 22: Effect of light level on contrast-detail measurements

Photon Densities in Fluoroscopy
(A – Chest; B – AP abdomen; C- LAT abdomen)
1 – Patient Entrance – identical for all three.
2 – Patient Exit
3 – Absorbed by Screen
4 – Light emitted by Screen
5 – Light entering Eye
6 – Absorbed by Retina

Source - Sturm & Morgan 1949[15] permission pending



Photon Densities



Corresponding Experimental Contrast-Detail Measurements,

VII. IMAGE INTENSIFIER WITH ANALOG IMAGE HANDLING (1955-1985)

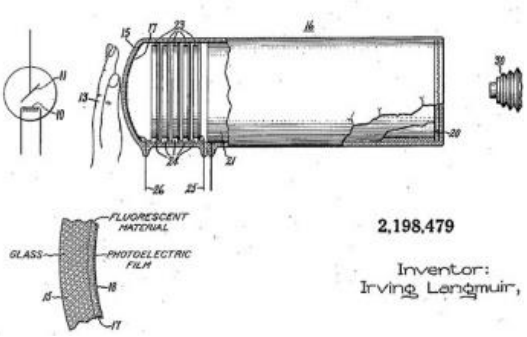
Visibility of objects is influenced by brightness, contrast, object size, and visual noise. Chamberlain's 1942 Carman lecture [13] provided major insights into the influence of the human visual system on fluoroscopic imaging and introduced the first technical requirements for fluoroscopic image intensification. The enormous operating range of the human eye (Fig 16) is remarkable for any physical sensor. The eye's ability to discriminate differences is much lower when the rods are the active elements (low-level scotopic vision) compared to the cones (high light level photopic vision) (Fig 17,19). In 1948 Rose described the sensitivity of the human eye on an absolute scale [16] He summarized his finding as "The performance of the eye over the bulk of its operating range may be matched by an ideal picture pickup device having a storage time of 0.2 second and a quantum efficiency of 5 percent at low lights decreasing to 0.5 percent at high lights".

Chamberlain estimated that the brightness of a fluoroscopic image had to be increased by a factor of a thousand to enable photopic vision. This paper included a 1940 patent drawing from GE of an image intensifier (Fig 23). Several patent drawings from this time-period claim the image intensifier as a radiographic device instead of for use as a fluoroscopic tool.

Fig 23: 1940 Image Intensifier Patient Drawing

This patent drawing illustrates all the essential elements of an image intensifier except for minification (the output image is the same size as the input image.

Source - U.S. Patent 2,198,479



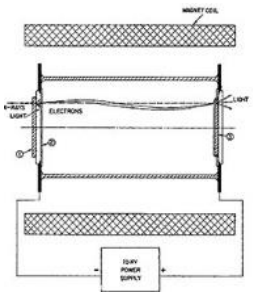
Coltman's 1948 paper 'Fluoroscopic Image Brightening by Electronic Means' [17] fuses the concepts of amplification and minification into what became the typical design of the fluoroscopic image intensifier. He also introduced a consideration of the effects of X-ray quantum limits on the perception of low-dose images. One important quote from this paper: "this is the fundamental limitation in image amplification, and that any system which does not make the fullest possible use of the available X-ray quanta incurs a deterioration of the image which cannot be corrected by subsequent amplification".

Figures 24 and 25 reproduces Coltman's electronic amplification only prototype device (perhaps based on a WW2 low-light goggle) and a sketch of an image intensifier tube with minification. Clinical X-ray intensifiers incorporate electro-optical minification (perhaps inspired by optical minification in photofluorographic chest systems), and electronic amplification. Image minification greatly improves the output light collection efficiency from such a tube. Figure 26 diagrams the conversion stages in an image intensifier tube that uses both electronic amplification and minification gain.

Fig 24: 1948 Prototype Image Intensifier

This device has electronic gain but does not utilize minification (input and output are the same size)

Source - Coltman 1948[17] – permission pending



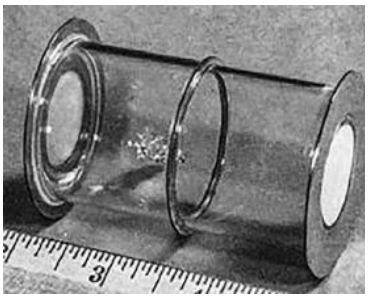
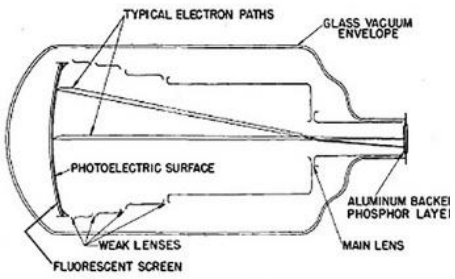
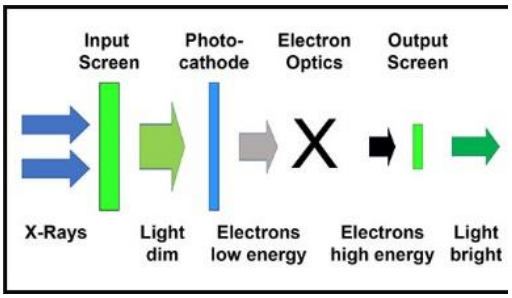


Fig 25: Image intensifier with minification



a) Schematic Source - Coltman 1948[17] – permission pending



b) Conversion Stages

Image intensifiers produce a much brighter image than a fluoroscopic screen for the same X-ray input dose rate. In principle, the measurement is simply measuring the light output of a 'standard' fluoroscopic screen and the image intensifier under the same irradiation conditions. Practical problems include the definition of a 'standard' screen, and the performance of light meters in the 1960s. A standard based on the ratio of light out of the image intensifier divided by the X-ray input (Gx) was developed by the ICRU and published as NBS handbook 89 in 1963[18]. The measurement of Gx was reviewed in detail by Holm and Mosely in 1964[19]. This paper reports measurements on a single image intensifier and discusses the effects of X-ray spectral differences on the measurements and changes in Gx over time attributed to deterioration of the input layers of the tube.

Figure 26 schematically illustrates the measurement process and the effects of collimation and zoom on a tube's conversion factor. An X-ray beam having standard characteristics illuminates the tube's entrance. An ion chamber is used to

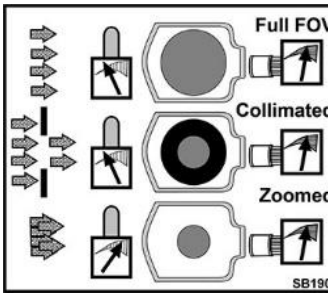
measure exposure-rate in the entrance plane and in the center of the field. A light meter measures light output (Cd/m^2) in the center of the output image. The Full FOV sketch indicates relative reference conditions. Neither measurement changes when the beam is collimated within the full FOV. However, the tube's minification gain diminishes when the tube is zoomed. More radiation is needed to produce the same light output. For an image intensifier: the radiation level scales with the ratio of the areas of the input and output screens (e.g. as the square of the ratio of input and output screen diameters). Looking ahead by four decades: In terms of radiation levels, zooming a flat-panel detector uses more or less of its input surface (similar to collimating an II). There is no need to adjust radiation levels with FOV because the brightness gain is independent of FOV. (Compensating for perceived noise as a function of FP zoom is discussed below.)

Fig 26: Image intensifier Conversion Factor
Gx = Light out / X-ray in. Gx is relative in this figure.

a) Max FOV – Low radiation to achieve light level: Gx = 10

b) Beam collimated within max FOV – Same radiation and light: Gx =10

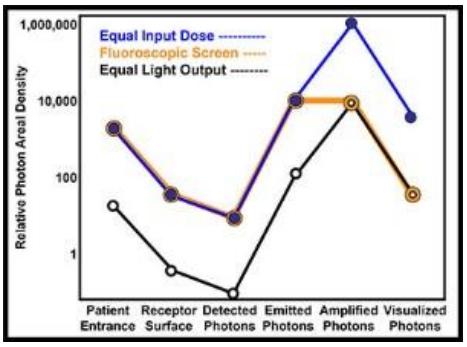
c) Image intensifier zoomed – Higher dose rate needed to achieve the same light level: Gx-5



Coltman's assertion that amplification alone cannot be used to decrease irradiation is demonstrated with the use of a quantum flux diagram (Figure 27). This diagram adds an additional stage for light amplification to the stages in Fig 21. Trajectory A is for a fluoroscopic screen and is identical to the quantum limited trajectory in Figure 21 (amplification factor is 1.0). Trajectory B has the same input dose rate as A and a light amplification factor of 100, about enough to enable photopic vision. For trajectory C, the patient input dose rate is reduced by 100 to produce the same output light level as the screen (A). Now, the quantum sink is two orders of magnitude lower, producing an image that is too noisy for clinical use.

Fig 27: Image intensifier photon-flux diagram

See text

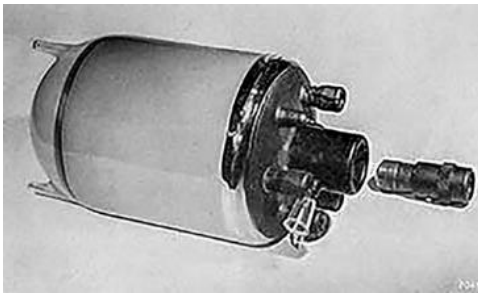


Coltman recommended constructing a system with a five-inch input screen based on considerations of clinical field-of-view, and bulk of the system. Its length was determined by electron-optics and scales with the size of the input screen. He stated: "The entire tube together with its housing, optical system, and protective lead shields, will be light enough to mount in place of the present fluoroscopic screen assembly on existing equipment". Figure 28 illustrates a different but representative 1955 single mode 5-inch tube.

Fig 28: c 1955 5" Image Intensifier

Note the optical relay lens which sent all available light from the II to the monocular viewing optic used by the system

Source - Philips Healthcare



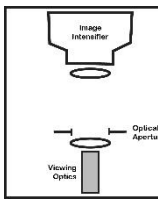
With image intensifiers, dark adaptation and dark fluoroscopic rooms were no longer required. Early image intensifiers had brightness gains of a few hundred times that of the brightness of the input screen. To permit operation at normal room

light levels, as many of the light photons produced at the output as possible had to reach the observer's retina. Monocular coupling was provided using optical periscopes. Figure 29 illustrates this in a 1956 Philips BV-20 mobile image intensifier with monocular viewing. A schematic drawing of the optical elements of this system is also shown.

Fig 29: 1956 Mobile Image Intensifier

a) Source - Philips Healthcare
b) Viewing - Photographed 2017
c) Monocular periscope output
d) Sketch of optics



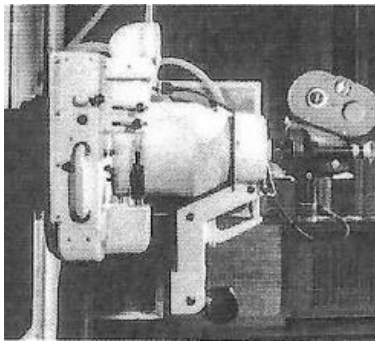
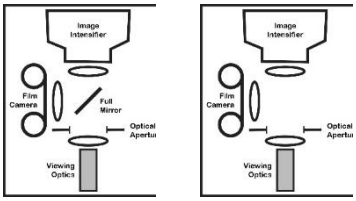



Another variant of a single channel system is shown in Fig 30. Figure 30 illustrates a c 1955 table mounted gantry equipped with monocular viewing, a photographic camera that could be placed into the single viewing channel. The operator could not see the images while filming. Its optics are shown schematically for fluoroscopy and fluorography.

Fig 30: 1956 Cine image intensifier

a) single output channel for either monocular periscope fluoroscopy or cine-fluorography
Source - Philips Healthcare

b) Sketch of optics in fluoro and cine modes

By the mid to late 1960s, direct fluoroscopic screens were increasingly replaced by image intensifiers. Figure 31 illustrates an image-intensified fluoroscope offered on the GE exhibit at the 1966 RSNA (my photo). The table includes a spot-film-tower mounted image intensifier. Looking at the right side of the photo, a direct fluoroscopic screen version was also offered for this system. Direct fluoroscopic screens remained in service throughout the world until their use was locally banned by local regulatory authorities. New direct screen fluoroscopic systems, including models with cryptoscopes, are still advertised on the internet. Direct screens may still be appropriate in regions where money is scarce and equipment service availability is low.

Fig 31: Transition

a) 1966 RSNA: System offered with either an image intensifier or a fluoroscopic screen

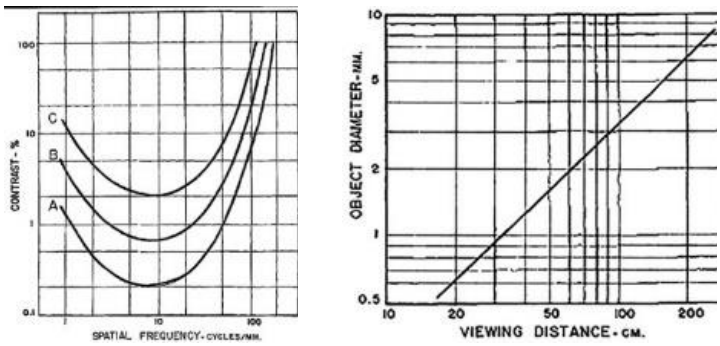
b) 2013 System with a fluoroscopic screen offered for sale on the internet.

In 1966 Morgan reviewed fluoroscopic imaging requirements in conjunction with the spatial and temporal performance of the observer's eye and available video technologies [20]. Figure 32 summarizes his findings for imaging a fixed object and a relatively high light level. This figure demonstrates both the effects of retinal physiology on the visibility of large or small objects and the influence of image noise (radiation dose) on minimum detectable contrast. The angle that an object subtends on the retina depends on both its size in the image and the observer's distance from the image. Higher dose levels are needed to suppress the noise conspicuity on more highly magnified images. A similar effect can occur when the large format viewing monitors found in many modern systems provides secondary image magnification.

Fig 32: 1966 Perception

"Threshold contrast plotted as a function of spatial frequency recorded on an observer's retina for three levels of noise contrast. Location of curves is shifted progressively upward as noise contrast increases."

Source- Morgan 66[20]– permission pending



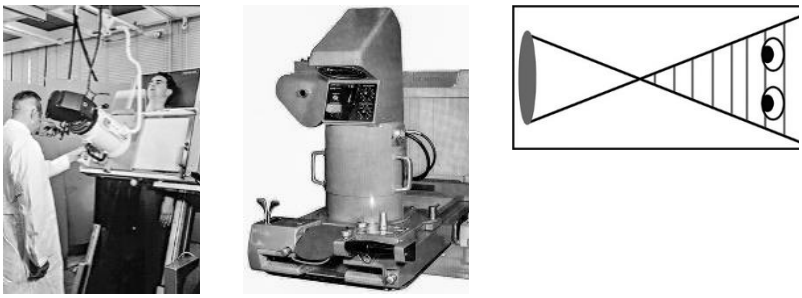
Improved image intensifier photocathodes increased the brightness gain into the few-thousand range. This yielded enough light to replace the periscope with mirror optics. Figure 33 illustrates two typical mirror-optic systems. Even though most of the light photons produced by the system did not enter the observer's eyes, there were enough photons on the retina to limit X-ray quantum noise to an acceptable level without the need for excessive patient dose. As predicted by Morgan [20], noise perception using systems with two or more fields-of-view, increasing magnification by selecting a smaller field-of-view required an increase in radiation dose for equal noise perception. This, along with the associated loss of minification gain in the image intensifier lead to dose-rate increases proportional to the ratio of the areas of the larger and smaller input FOVs. The interactions of radiation-dose and noise perception as a function of FOV remain important today for perceptual reasons.

Fig 33: Mirror Viewing

a) c 1960 Note backup direct fluoroscopic screen Source - Philips Healthcare

b) 1964 Note slot for spot-film cassette Source - PICKER advertisement in RADIOLOGY

c) Exit Pupil: Must be large enough to illuminate both of the observer's eyes. Most of the light from the II does not enter the eye.



Collimating the X-ray beam to the clinical area of interest is an important radioprotetive measure for both patients and staff. Collimation also improves object visibility by reducing the scatter contribution to the image. The need for collimation was recognized by 1905 (Fig 7, 9), but often not provided (Fig 10,15) The introduction of small round image intensifiers posed a second issue (Fig 33). Collimators also had to irradiate the larger rectangular films used in spot-film devices. Rectangular collimators were supplied to meet filming requirements. There will be irradiated but not imaged areas in the patient when a square or rectangular collimator is used to fully illuminate the input of a round image intensifier.

Figure 34 shows Dr. F. Mason Sones performing diagnostic cardiac angiography using a 1956 11-inch image intensifier equipped with cinefluorography and periscopic viewing. The imaging system was too bulky to place over the patient. Thus, the imaging system, and operator, were in a pit below the patient table. A second individual handled the intravascular catheters without any ability to see the fluoroscopic images. This system has a beam splitter that permits viewing while filming. It also includes a phototube that measures the light intensity in a defined portion of the image (measuring field). A schematic of the optics of functionally similar later system is also shown. A later photograph of such a 'pit' lab includes an additional 5-inch over-table image intensifier with mirror viewing optics

Fig 34: Early Cardiac II

a) The 11" II was too bulky to operate above the patient.(1955).

b)1960 photo of a 'pit' lab shows a 5" over table II with mirror optics added to the setup .

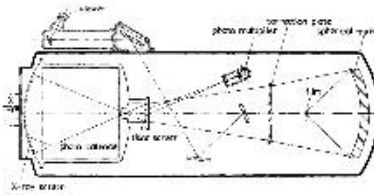
c) Schematic sketch of optics

(a-c) Source - Philips Healthcare

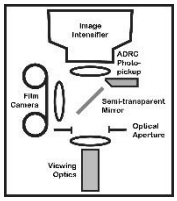
d) corresponding optics sketch

A

B



C



D

As noted above, this 11-inch device includes a photo pickup which indicates that the fluoroscopic system included an Automatic Brightness Control (ABC) circuit. Because such circuits work by controlling radiation output, the nomenclature changed to Automatic Dose Rate Control (ADRC). Photo pickups were incorporated into the optics used for most of the systems incorporating analog video chains. Further examples are shown in the sketches of image intensifier optics presented later in this paper.

Photo pickups measure the average light intensity in a defined area of the image intensifier's output image. X-ray factors are increased if the measured light level is too low and decreased if it is too high. The controlled elements were a combination of kVp, mA, and fluoroscopic pulse-width. The overall image became too bright if portions of the image in the measuring field included large amounts of contrast medium such as barium or iodine. Some systems provided a 'lock' option which allowed the operator to turn off the ADRC just before administering contrast. Investigators researching video densitometry in the 1970s and 1980s [21, 22] obtained unusual results if they did not understand the relationship between ADRC and contrast in the imaging field. Once images were digitized, ADRC control utilized analysis of appropriate areas of the digital image. Controls could now account for collimator size and position, as well as using examination specific measuring fields.

In many systems, the fluoroscopic ADRC sets system parameters for the next acquisition. If a different projection angle or body part is involved, it is a good idea to fluoroscope for a second before starting an acquisition.

Multi-mode (zoom) image intensifiers contain sets of internal electrodes which focus the photoelectrons collected from all or a portion of the photocathode onto a fixed output screen (Figure 35). This form of electronic magnification the system's required increase input radiation dose rates and could increase overall high-contrast resolution if the output-screen or other output elements are the resolution limiters. The minification gain of an image intensifier is proportional to the ratio of the areas of the input and output screens. To maintain constant brightness, the input dose rate increases inversely as the square of the diameter of the input screen.

Fig 35: c 1975

a) 9-inch dual field (6-9) image intensifier

b) Image intensifier factory
Vacuum pumping and outgassing stations are shown.

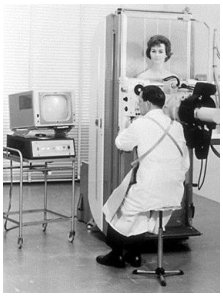
Source - Philips Healthcare

Analog video was introduced in the 1960s. This development removed a major limitation on the position of the operator relative to the patient (FIG 36). As a bonus, everyone in the room (including the patient) could see the fluoroscopic images It also enabled the development of remote-control fluoroscopes with the image intensifier under the table (Figure 37). Placing the image receptor under the table results in having the X-ray tube over the table. The irradiation of an operator's eyes is

potentially increased from backscatter from the X-ray beam's patient entrance port in this geometry. To minimize staff head irradiation, most modern fluoroscopic systems place the image receptor above the patient.

Fig 36: Video Fluoroscopy c 1975

a,b) Note both conventional film/screen and fluorographic acquisition devices
Source = Philips Healthcare
c) Sketch of optics with beam splitter for acquisition



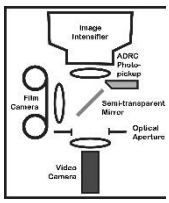
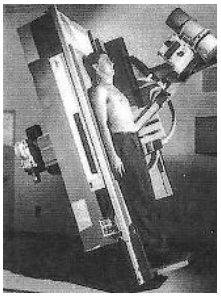


Fig 37: c 1978 Remote Control systems with under- table image intensifiers.

Increased above table scatter intensity may be a hazard.

a) Philips Healthcare
b) Siemens Healthineers
c) German Post Office

A



B

C

Fluoroscopic video systems usually used the broadcast video formats and standards of the countries where the systems were installed. The European standard was 25 frames per second with 625 lines per frame. The corresponding United States standard was 30 frames per second with 525 lines per frame. Both systems used 2:1 interlaced video to reduce image flicker. These values were selected to use and synchronize with the cycle rate of public power distribution systems (50 Hz in Europe; 60 Hz in the USA). Higher line densities and higher frame rates were well known in this era but only occasionally found in clinical systems.

The USA and European analog video systems were incompatible. Analog scan converters were developed to facilitate trans-border movement of broadcast video streams such as the Olympics. These devices essentially 'wrote' charge distributions representing the image onto a storage mesh while simultaneously the mesh was scanned using an electron beam in the other format. For visual reasons, neither did a single frame write saturate the mesh nor did a single frame read discharge the mesh. The analog scan converter tube was adapted as an image integration and storage element in both fluoroscopy and B-scan ultrasonic imaging. Analog scan-converters were eventually succeeded by digital counterparts. Digital-scan-converters are currently used for a variety of medical and non-medical applications. Image characteristics of video-based systems are discussed later in this paper.

Even early image intensifier systems provided enough light to expose photographic film using acceptable patient dose levels using doses per frame about ten times higher than fluoroscopy. To meet the greater image quality requirements in fluorography relative to fluoroscopy, this ratio is still broadly valid in the digital world of 2019. The image intensifier is the enabling technology for photo-fluorography (photographing single or slow sequences of images directly from the image intensifier) and cine-fluorography (photography at motion picture frame rates). Film-screen based spot-films continued in use for several decades for tasks such as gastrointestinal fluoroscopy. Cinefluorography (Cine) was rapidly adopted for imaging rapidly moving structures such as the heart. Figure 38 illustrates a 23 cm (9 inch) image intensifier equipped with a cine camera and back-up mirror optics. This system also has a television camera.

Fig 38: c 1970 Cine + video

Video and cine with a back-up mirror optic

Source – Philips Healthcare

Reliable analog video cameras suitable for routine hospital use were available by the late 1960s. These cameras eliminated the need for even back-up optical viewing. Video viewing became a major enabler of fluoroscopically guided angiography because video removed major constraints on the operator's position and posture. Figure 39 was taken in Dr. Sones' lab after video was introduced a few years later. The operator is now able to work at tableside to manage the patient and manipulate the angiographic catheter directly. Others in the room can also see the fluoroscopic images as they work. A similar c 1980 cardiac catheterization laboratory is shown on the right. The X-ray beam path is vertical both systems. Patients were rotated to achieve necessary clinical projections. Note the patient cradle in the 1980 laboratory.

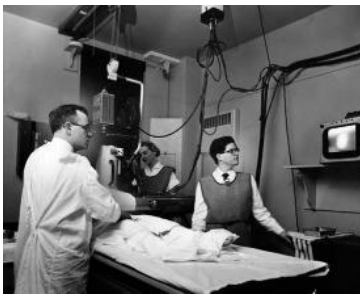
Among other things, the use of video-fluorography[23] as a substitute for film placed in front of the image intensifier reopened the beam collimation question. An iris collimator that could provide a circular beam adjustable for both anatomic interest and the source-to-receptor distance (SID) is an ideal first step. Systems with rectangular collimators could be set up to fully irradiate the image intensifier (important for small FOVs), fully enclose the beam within the II's FOV, or provide some balance. Regulatory limits allowed necessary over-irradiation plus a small extra margin to accommodate mechanical tolerances. The latest draft equipment standards recognize that this is no longer necessary and will require the beam to be confined to within the active FOV.

Fig 39: Cardiac Video Viewing

X-ray system has a fixed vertical axis in both images.

Patient ls rotated for different clinical views in system b

Source - Philips Healthcare



Gantries with one and two degrees of rotational freedom were introduced in the 1970s (Figure 40). Moving gantries were safer for patents but needed new forms of staff scatter protection to replace that provided by fixed X-Ray tubes located in a shielded table base. Simultaneously optimizing patient safety, physician-patient access, and staff safety is still evolving. There is an extensive literature on this topic. A small subset is referenced in this paper [24-41].

Fig 40: Rotating X-ray gantries

Moving gantries introduced new concerns about operator irradiation

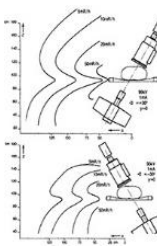
a) Scatter fields for system B

b) 1975 One rotational axis

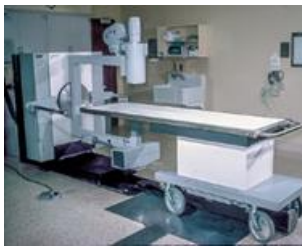
Source - Philips Healthcare

c) 1995 Two rotational axis

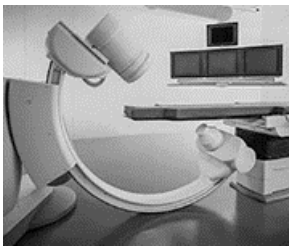
Source - Siemens Healthineers



A



B



C

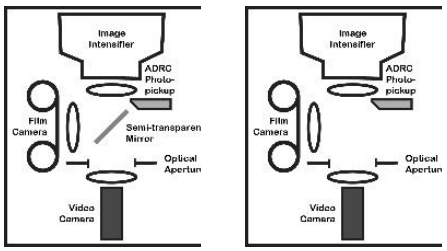
Most image intensifier assemblies used visible light to couple the output of the image intensifier tube to the observer's eye and/or cameras. Sketches of the optical handling systems (distributor) were shown in previous figures and continue in Figures 41 and 42.

Key elements in the distributor shown in Figures 41 and 42 include the photo pickup tube used by the ADRC, an optical diaphragm, and beam-splitters. In systems corresponding to Figure 40, the diaphragm was typically partially closed (degrading system brightness gain) to force enough X-ray flux to limit fluoroscopic X-ray quantum noise. Film cameras physically require more light than video cameras or the eye. The ADRC was programmed to accommodate that need resulting in less-noisy acquisition images. The beam-splitter diverted about 10% of the light into the viewing channel during photofluorography, allowing simultaneous viewing and image acquisition. The beam-splitter was removed during fluoroscopy, all available light was sent to the viewing channel, reducing X-ray dose rates to about 10% of photofluorographic levels.

Fig 41: Optical distributer with video and fluorographic cameras.

A 90%-10% beam splitter is used for fluorography which sends most of the light to the film-camera and passes enough to operate the video camera.

The beam splitter is mechanically removed for fluoroscopy to send all available light to video camera and thus minimize dose-rate.



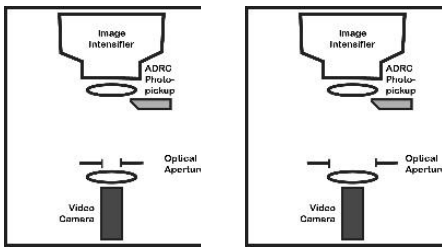
A - Cine **B - Fluoro**

The aperture had an additional function when photofluorography was replaced by video-fluorography (Figure 42). Closing the aperture for fluorography reduces the effective brightness gain of the system. This results in a higher x-ray dose rate for fluorography while maintaining a similar illumination level of the video chain for fluoroscopy and fluorography. Because the light level at the camera was similar for each mode, camera noise levels were similar.

Fig 42: Optical distributer with video-fluorography and optical ADRC

Optical aperture is closed for fluorography to minimize image noise. This requires an increased dose-rate for fluorography

Optical aperture is opened for fluoroscopy to minimize dose-rate



A - Cine **B - Fluoro**

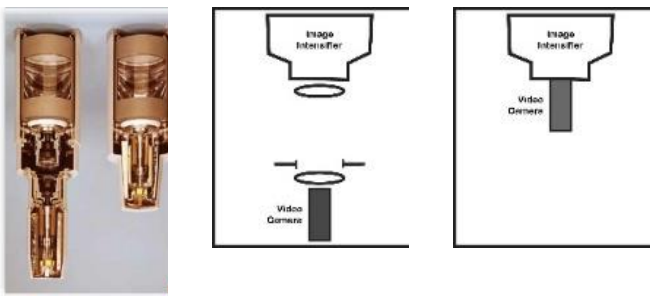
The separate ADRC optical sensor was replaced by defining a ROI on the video images. The controlled optical aperture was retained for cameras with limited dynamic range. The aperture could be removed once wide dynamic cameras became available. Eventually the camera was placed in physical contact with the output of the image intensifier (Figure 43). X-ray and camera factors were set by selecting the operating mode of the system. In these configurations, camera noise is relatively higher for fluoroscopy than for video fluorography.

Fig 43: Optical distributer with video-fluorography and video ADRC

Optical aperture is opened for fluoroscopy to minimize dose-rate Optical aperture is closed for fluorography to minimize image noise. This requires an increased dose-rate

The entire optical system can be eliminated once video cameras with enough dynamic range became available. The dose-rate and pick-up camera operating conditions are set by selecting the fluoroscope's operating mode (fluoroscopy or acquisition).

Source - Philips Healthcare



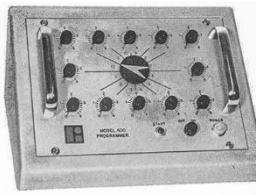
Film recording of image intensifier output images (fluorography) did not meet many non-cine clinical requirements until the 1970s. For example: Small FOV image intensifiers cannot image large anatomical structures without panning the imager. Direct film-screen systems (either spot-film devices or rapid film changers) were used and further developed in the 1960s.

Figure 44 illustrates typical film changers used for angiography. The fluoroscopic tube and its image intensifier were often separate assembly. Once the angiographic catheter was placed, the region of interest was moved to the film-changer zone (in some systems, the film-changer mechanically replaced the image-intensifier. Film-changers had maximum frame rates of 6 per second, and very limited film capacity (approx. 10 – 50 images). To complete a study with limited image capability, film-changers included control elements that varied framerates during a single angiographic run. Typically, there was a programmable time delay from the start of contrast-media injection until the first images; relatively rapid imaging during the arterial phase of the study (2-4 images per second) and slower imaging (approx. 1 image per second) during the wash-out and venous phases. Programmable frame timing saves radiation as well as film. Variable acquisition frame rates are still available on many c 2010 – 2020 digital angiographic systems. They are a valuable radiation management tool.

Fig 44: rapid film changer and controller

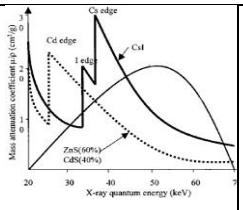
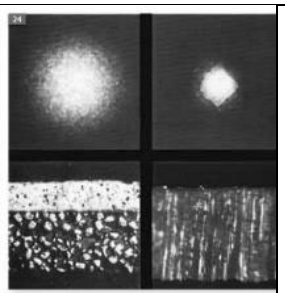
a) The AP changer is under the table at the left, the lateral changer is away from the table. The image intensifier is in the center. Separate X-ray tubes are used with each. Source - 1974 St. Vincent Hospital, Worcester MA

b) Sanchez-Perez cassette changer control module. Capable of programming variable frame rates during a single acquisition run. Source – Radiology Advertisement 1964



Until the mid-1970s, image intensifiers used the same ZnCdS phosphor for the X-ray detection as did the direct fluoroscopic screen. ZnCdS was deployed as small crystals dispersed in a binder (Figure 45). A fluoroscopic screen's efficiency in converting X-rays into visible light is proportional to the fraction of X-ray photons absorbed with active crystals (stopping fraction) and the crystal's ability to convert X-ray energy into visible light (conversion factor). Increasing the stopping fraction of a screen is accomplished by increasing the thickness of the screen. Thicker screens give light photons more room to scatter before they encounter the detector layer (the photocathode in the case of an image intensifier). Light scatter decreases image sharpness. The best image intensifiers had a stopping power of about 1/3. Patient dose rates were not very different for these two technologies under optimum conditions.

The introduction of CsI as a replacement input phosphor simultaneously reduced dose requirements by about a factor of two because of the higher X-ray stopping power of CsI, greater packing density of the active phosphor and reduced optical scattering in the screen because its structure is composed of crystals acting as light pipes. Figure 45 illustrates the differences between these two screens.

Fig 45: CsI Input Screen			
	A - Physics	**B -** Comparison of ZnCdS & CsI Source-Philips Healthcare	**C** – Machlett Advertisement in Radiology

VIII. ANALOG VIDEO (1960-2000)

Every available analog video-camera technology was employed at some time in fluoroscopic systems (Fig. 46). Major species included the image-orthicon and the vidicon. Orthicons were the usual broadcast-television pick-up tubes of the 1860S. Broadcast video standards were developed around the orthicon's bandwidth and characteristically minimum signal integration at the detector which minimized motion blur of live video. The image tubes themselves, along with the array of analog vacuum tubes in the camera were bulky and required a great deal of ongoing maintenance. The vidicon provided

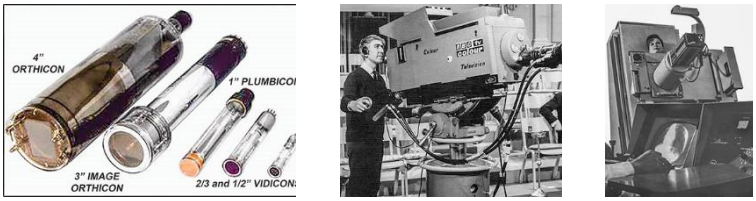
similar video outputs in a smaller package at the expense of increased image integration at the detector. The plumbicon is a form of vidicon, using lead-oxide in its detector, which exhibited less lag.

Fig 46: Video Tubes and cameras

a) Analog video tubes ranging from a 4" diameter orthicon to a ½" consumer vidicon

b) Orthicon broadcast camera Source – BBC

c) Orthicon Medical Camera Source Westinghouse ad in RADIOLOGY



A B C

Lag in a video tube provided a benefit for fluoroscopic imaging before the era of digital video image processing. When the lag was not excessive, the tube effectively temporally averaged X-ray quantum noise producing a smoother image. Surviving anecdotes from that era include the report that orthicons were tested for lag before installation in broadcast cameras. Some of these broadcast rejects provided better fluoroscopic visual performance than the accepted tubes because of noise integration. X-ray manufacturers were reported to advertise these tubes as 'selected for medical use'. I was personally involved in the installation of a fluoroscopic system for GI examinations featuring the "new" low-lag plumbicon. The radiologists using this system were quite upset at the visual noise in the images. Because this was before the era of real-time image processing, the plumbicon was replaced with a higher image-lag vidicon to meet the radiologist's clinical requirements. Today, digital fluoroscopes artificially reproduce camera-lag using recursive filtering in their image processing software.

Analog broadcast video was transmitted using interlaced scans as a means of increasing the frame rate while constraining bandwidth. This is beneficial because the increased rate decreases the impression of flicker. Commercial motion pictures achieved the same result by projecting the same frame twice before advancing the film. Video interlacing results in a time lag between the two fields comprising a video frame (e.g. 1/60 of a second for 30 fps). Some anatomical structures (e.g. right coronary artery at systole) move fast enough to cause the structure's image to double. High frame rate, high bandwidth, non-interlaced video systems were available. While they performed well, the lack of standardization along with limited compatible third-party hardware, made image transport extremely difficult.

Analog video sequences can be recorded using either analog videotape or analog video disk. Maintaining image quality on these devices was very labor intensive. By the late 1970s, analog to digital scan-converters were able to extract either a single digital image or a sequence of digital images. These images were then locally stored digitally for processing and/or viewing. The availability of electronic images provided the technical infrastructure needed for digital angiography and digital subtraction angiography. The technical resources (networks, media, and displays) needed for digital archives were not available in that era. Digital fluorographic images were recorded on film for archiving using external laser or CRT based cameras.

IX. IMAGE INTENSIFIERS WITH DIGITAL IMAGE HANDLING and X-RAY BEAM SPECTRAL MANAGEMENT (1980-2000)

Fluoroscopic image outputs moved from pure analog imaging chains to hybrids with a single digital output for both fluoroscopy and fluorography in the 1980s. This was initially accomplished by processing the analog video signal using an analog to digital converter. Eventually a CCD camera replaced the analog video pickup tube. Hardware limitations early in this era constrained the output digital images to a nominal 512 x 512 pixel matrix with a maximum depth of 8 bits. Such images are adequate if detail averaging in a single pixel is not excessive and the intensity range in the entire image is accommodated within the available bit depth. Analog video – converter box systems were replaced by digital cameras (e.g. an array of digital sensor elements) with similar specifications. Technological advanced allowed greater matrix sizes and bit depts., while maintaining frame rates exceeding 30 fps.

By 1990, imaging chain design had reached the point where X-ray quantum noise was the major noise source for fluoroscopy. There was less than a factor of two available for reducing patient irradiation by improving the image intensifier. Radiation levels can be further reduced if radiological conspicuity of clinically important items, such as contrast media and guide wires, could be increased.

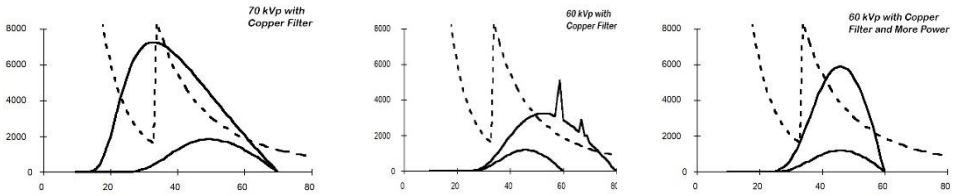
One way of improving the conspicuity of higher Z elements is to maximize the fraction of X-Ray photons in the beam with energies slightly above the K absorption edge of the material of interest. (e.g. iodine at 33.179 KeV). Filtering the beam by adding a layer of copper (0.1 – 1.0 mm) can accomplish this goal [42, 43]. Figure 47 is a sketch of the effect of this filter for a typical 70 KVp spectrum. Adding copper greatly reduces the fraction of the beam with energies below the iodine edge. It also decreases the total photon flux available to form the image. Increasing the operating voltage to 80 KVp does not help because most of the flux gain occurs at photon energies substantially higher than the iodine K edge. Reducing the KVP to 60

and increasing the tube current moves the photon spectral distribution closer to the Iodine K edge while simultaneously providing enough flux to meet imaging statistical requirements. This option became available in the 1990s when X-ray tubes were developed with enough long-time power ratings to meet this requirement without damage (Figure 48). The increased anode cooling in this class of tubes is partially attributable to the replacement of ball-bearing support for the anode assembly with liquid bearings.

Fig 47 Beam spectral shaping

See text

Dotted line is Iodine attenuation

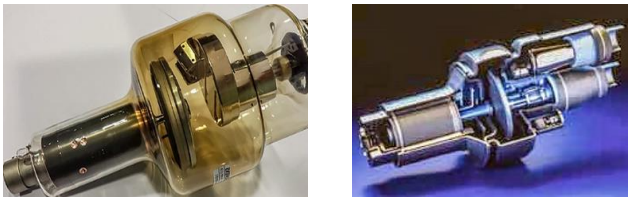


A – 70 kV with and without Cu **B** – 60,80 kV with Cu **C –** 60 kV with increased mA

Fig 48: X-ray tube evolution

a) c 1980 – Primarily radiative cooling Source- Siemens Healthineers

b) c 1990 Conduction cooling via liquid metal bearings and external heat exchanger Source- Philips Healthcare



X. DIGITAL IMAGE PROCESSING (1980-2010)

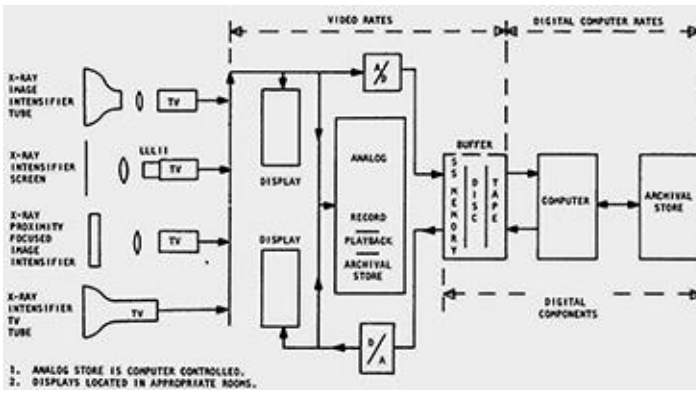
Electronic processing of clinical radiographic and fluoroscopic images began in the late 1970's. Capp's 1980 New Horizons lecture at the RSNA[44] provides a snapshot of this watershed (Fig 49). The transition from analog to digital video was much earlier than Capp's forecast. It was driven by increasing availability of small computers capable of image processing at video speeds (e.g. 30 fps) and the replacement of analog video-tubes with CCD cameras. Two major results were real-time processing of fluoroscopic imaging streams and Digital Subtraction Angiography (DSA)

Managing the characteristics of analog video images was challenging. In this era, temporal noise averaging was physically achieved by selecting video pickup tubes and display monitors with appropriate characteristics. All processing affected the entire image. Overall temporal resolution was determined by the video-framerate, a value usually linked to the local power-line frequency for broadcast compatibility and synchronization. As discussed above, greater temporal integration reduced the appearance of quantum noise at the expense of increased motion blur. Image brightness and contrast could be electronically adjusted. Spatial resolution was strongly influenced by the scan pattern and number of available scan lines. None of these factors could be easily changed to either enhance imaging appearance in real time (e.g. automatic contrast adjustments) or optimize imaging to meet differing requirements of different procedures.

Fig 49: 1980 forecast of electronic image handling in 2000

In this view, image capture and image processing are in the analog domain. Image storage is digital. Film has disappeared.

Source – Capp[44] – permission pending



XI. DIGITAL SUBTRACTION ANGIOGRAPHY (DSA) – (1980-2019)

Photographic subtraction was used in the mechanical film-changer era to document blood flow. An image obtained before contrast-media reached the region of interest is selected as the mask. A negative version of this image is produced in the darkroom. The single negative mask was then physically registered with subsequent images to produce the subtracted image.

Unsharp masking involved placing a spacing sheet of glass between the mask and a later image. Mask blur is determined by the thickness of the spacer. This tedious procedure was seldom used to subtract more than a few of the images acquired during a study. Electronic subtraction in real-time or near-real-time was developed and deployed as soon as the enabling analog and technologies permitted [45-48]. Fig 50 provides system schematics of subtraction radiography and subtraction fluoroscopy. Much of the work is performed in the analog domain because the technology of this time did not support fully digital equipment. Figure 51 illustrates some of the hardware used for two early digital subtraction systems. The subtracted images were in video format and either recorded as video or photographed from a CRT display. Maintaining image quality for fluoroscopes that use image intensifiers and perform analog image processing is difficult because of instabilities and image noise added by these elements. The introduction of fully digital image processors enabled useful functions such as better registration of live and mask images, and road-mapping (partial subtraction that provided enough anatomical reference to facilitate device movement). DSA, including its sub modes, rapidly became, and remains, a cornerstone of FGI.

Fig 50: 1980 Hybrid Subtraction Angiography

See text

Source - Ergun[45] – permission pending

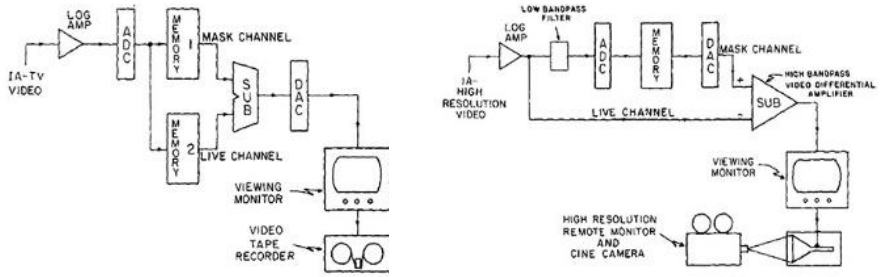
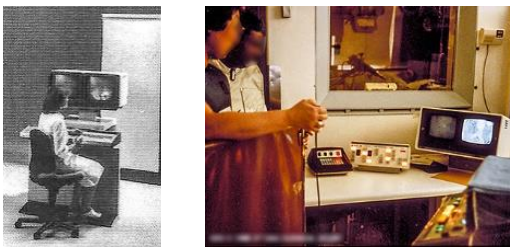


Fig 51: 1980 DSA Hardware

a) Note the size of the electronics cabinet needed to house the early `image processor. Source – Siemens Healthineers

b) Controls used for a first-generation clinical DSA system. The image processor for this device also filled a full-size electronics cabinet.



Subtraction angiography (e.g. DSA) removes fixed information found in both the mask and live image from the resultant. All X-ray images contain stochastic quantum noise with the point-to-point noise intensity proportional to the square root of the local signal intensity. Other noise sources may be present with different spatial distributions. Random noise does not subtract but adds and produces more noise in the difference image than in either initial image. The final step in DSA is to greatly increase the contrast of the display to improve visibility of blood vessels or other moving structures. This also increases the visibility of image noise. The target-to-noise ratio can only be increased by increasing radiation dose. Thus, DSA requires about an order of more radiation per frame than unsubtracted imaging of the same object. Unnecessary irradiation is minimized if X-ray quantum noise is the dominant noise source in an image. This can be evaluated if the noise in the subtracted image differs from place to place (e.g. near and far from subtracted bone). This appearance is called a noise-print[49] and can serve as a quick quality test using clinical DSA images.

DSA using CO_2 as the contrast agent (available since the 1980's) can be of clinical benefit. Images are recorded while a CO_2 gas injection displaces blood. The optimum technique uses a higher KvP to reduce unwanted photoelectric attenuation in the patient's tissues, and image stitching to produce a composite image or image series of the entire artery by following the CO_2 bubble through the patient's arteries over time.

XII. IMAGE PROCESSING, DISPLAY, AND STORAGE (1990 – 2010)

Digital video eliminates many of the constraints inherent in powerline synchronized analog video. Working back from the display toward the imaging hardware: Displays are not constrained by the imager's pixel size or acquisition frame rate. Images are interpolated and scaled to transform the acquisition version into the matrix size used for display. These images are refreshed in the display at a rate well above the eye's critical flicker frequency. In its simplest form, the same data is shown multiple times until replaced by a new image. In addition to decoupling acquisition and display rates, this provides the functionality for pulsed fluoroscopy and acquisition (X-ray production during a fraction of the available frame time).

Frames can also be combined in an arbitrary manner. DSA image processing is inherent, including technologies that were impossible with analog video such as frame by frame pixel shifts to compensate for unwanted anatomical motion during a run. Noise reduction is obtained by recursive filtering: A fraction of the previous image is added to the newly acquired data.

Noise reduction increases as the number of previous acquisitions contribute. The downside is that the remnants of the older images produce ghosts and blur of moving objects. This effect is programmable and can be configured at tableside to match the immediate clinical imaging requirement. Compare this with achieving this balance by changing video-tube technologies in the analog era.

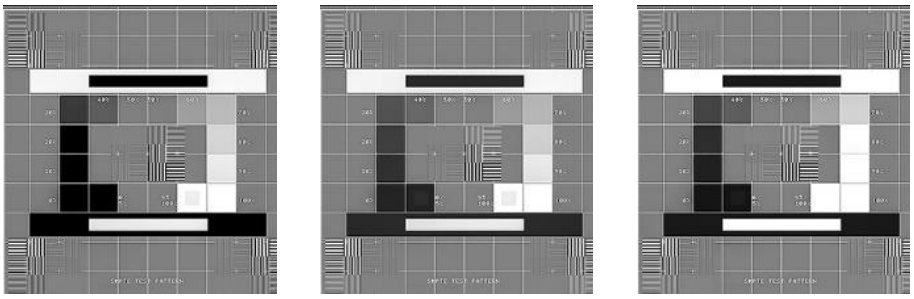
Images can be visually sharpened using a variety of processes such as unsharp masking. This technique combines sharp and blurred versions of the same image. A similar process occurs in the retina by interactions at a neural level. The results of the process can be seen as Mach bands outlining the edges of recorded or visualized objects. Here again, the magnitude of the effect in the digital domain can be configured at tableside.

The relationship between the X-ray intensity illuminating a detector pixel and the brightness of that pixel on a display is defined by an arbitrary look-up-table. A simple example is the window-level controls found on CT scanners as well as fluoroscopes. Different look-up-tables are applied to different imaging tasks, and each table can be immediately tuned as needed.

Viewing monitors are a non-trivial link in the fluoroscopic imaging chain. Monitors used for primary radiographic and mammographic diagnosis are available with built-in quality control tools. Current fluoroscopic systems often include an injected or stored digital SMPTE test pattern[50]. Objects are provided to assess spatial-resolution, gray-scale performance, as well as system generated artifacts. Figure 52 illustrates this test including examples of white and black clipping. Fluoroscopic monitors are expected to simultaneously show both the 5% block (in the 0% black block) and the 95% block (in the 100% white block) when viewed under clinical working laboratory light levels.

Fig 52: SMPTE Test Pattern
a) Black clipping
b) Full range – Expected with normal room light.
c) White clipping

Original Source - ANON



A **B** **C**

As described by Morgan in 1966 [20], the ability to perceive an object includes the effects such as the angle subtended by the object on the observer's retina. Increasing the angular size of an object by any means (e.g. using magnification mode in an image intensifier, digitally magnifying an image, or simply using a larger monitor) may affect its perceptibility. In some cases, reacting to larger images can improve operator radiation protection. Some interventional fluoroscopists using conventional (19") monitors tend to lean over the patients during imaging to better see small arteries. The same operators often stand erect when performing the same cases using large (60") monitors. They seem to be adjusting their posture to best see the target vessels. The posture change reduces their irradiation simply because it moves them away from the patient's scatter field.

Fig 53: Effect of monitor size on operator posture.
a) 19": The operator is leaning over the patient to better see the arteries during a cine run.

b) 60": The magnified image permits the operator to stand away from the patient.

Digital medical images are among the largest files (both individually and in total) in healthcare. In the context of this review, digital images can be stored within the fluoroscope or in an external archive such as a PACS. Image transport is via networks or media. The deployment of digital images was, and continues to be, facilitated by the evolving DICOM standard. A review of this area is beyond the space limitations of this paper.

Figure 54 provides a glimpse back into the transition era. These images were taken in the interventional cardiology laboratory of Lenox Hill Hospital in New York City. Cine media included film, analog videotape, and several digital CD formats.

Fig 54: c 2000 Transition from analog to digital image management

a) Cine film processor and darkroom.

b) Cine film viewer. The boxes in the background are part of the daily image transfer between the lab and its off-site archive. Each box contains 50 rolls of cinefilm. Image storage and retrieval was on a 24-hour cycle.

c) One day's cases for one physician. Images are on film, CD, and analog videotape.

Source for all - Balter – Lenox Hill Hospital, New York

XIII. EXAMINATION CONFIGURATION

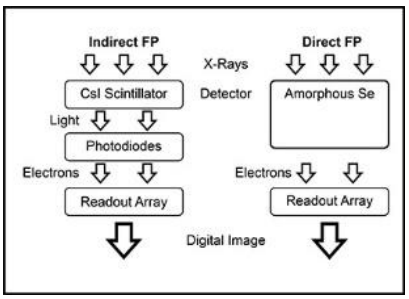
Fluoroscopic procedures usually include the operator's direct adjustments of imaging parameters and often substantial a-priori information about the specific patient's medical condition. Almost all the fluoroscopes manufactured after the 1970s provide a selection of pre-configured examination and patient specific technical sets. Many of the pre-configured X-ray production and image processing factors can be overridden by operators to suit their immediate imaging needs while a procedure is in progress. At present, high-end fluoroscopes may be set to any of several thousand distinct configurations. Each configuration potentially differs from the others in its radiation management behavior, its image processing characteristics, and usually in both domains.

XIV. SOLID-STATE FLUOROSCOPIC IMAGE RECEPTORS (2000-2019)

The image intensifier tubes themselves began to be replaced by solid-state detectors starting c 2000 [51-56]. Solid state detectors (flat panels or FP) were commercially available in 2000. Both indirect (X-ray to light to electronic signal) and direct (X-ray directly to electronic signal) technologies were known. (Figure 55)

Fig 55: Indirect and Direct Flat Panel Detectors

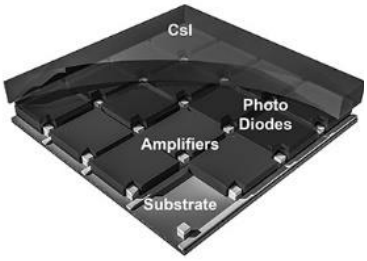
Schematic sketch of information carriers



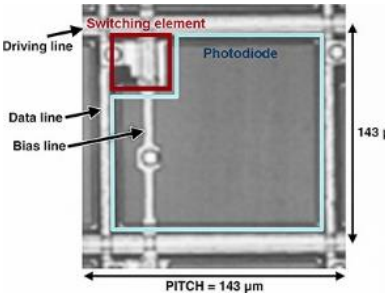
The same CsI scintillator is used in late-model image intensifiers and indirect FPS. Input dose rates for fluoroscopy and fluorography are similar in both domains. However electronic noise levels are somewhat higher in a FPS and amplified with digital magnification. This does not occur in an II using electro-optical zoom. Thus, fluoroscopic dose rates are usually to be a bit higher in FPS compared to late-model image intensifiers. Dose rates needed to meet the imaging requirements of fluorography are sufficiently high that they are independent of technology. Radiation savings attributable to FPS are due to better radiation management in newer designs[57].

The need for video frame rates (e.g. 30 fps) resulted in the use of indirect technology for the first generation of FP detectors. This technology continues to be found in most FP detectors produced in 2019. The radiation detection element that is continuous sheet of CsI, essentially identical to the CsI element in the image intensifier of the era. The detector's nominal physical size was 20 x 20 cm. Light from the CsI was detected by an array of 1000 x 1000 photodiodes/amplifiers. The nominal pixel size of this device was a bit under 0.2 x 0.2 mm. Space is needed in the photodiode array for amplifiers and readout electronics. Thus, the active surface of a single photodiode only filled about 70% of the nominal pixel size. X-ray photons interacting in other areas contributed to the signal due to light diffusion in the CsI layer. Figure 56 illustrates hardware elements of indirect FP detectors

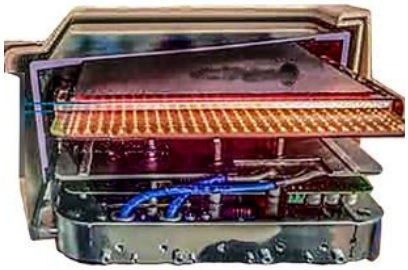
Fig 56: Indirect Flat Panel



Detector elements



Pixel – Source - Philips Healthcare



Assembly with electronics and thermal regulators. – Source - Philips Healthcare

FPs reduce or eliminate several inherent image-intensifier artifacts. These include vingetting (images dimmer in the periphery relative to the center), pincushion-distortion (result of projecting a curved input screen onto a flat output screen). S-distortion (electro-magnetic field influence on the II's electron optics), and II dynamic range limitations.

Three separate actions occur when the field of view of a flat-panel system is changed: 1) Irradiation is limited to a subset of the pixels in the detector. 2) Detector pixels may be binned to comply with bandwidth limitations of the detector's output channel, and 3) The output image is scaled by an external image processor to fit the properties of the image display. The spatial resolution of a FP varies with magnification mode in a fundamentally different way than for an image intensifier. In an II, the limiting resolution element is the output screen or video camera. Changing the FOV in an II projects a larger or smaller portion of the input screen onto the fixed size output system. In the II, resolution increases inversely with the diameter of the input FOV. However, in a FP, a greater or lesser number of fixed size pixels are irradiated when the FOV is changed. The inherent resolution of the detector elements is unaffected by changes in FOV.

Spatial resolution of a FP system is affected by two additional factors: In large format FPs, the total number of pixels exceeds the fluoroscope's image matrix size. Pixels are binned in the detector electronics to reduce the output matrix for large FOVs. (typically, 2 x 2 into 1 x 1). Binning is removed when the FOV is small enough to fit the output matrix. The resultant spatial resolution has a step increase at the FOV where binning is no longer applied. A second, less common, use of binning is to accommodate very fast video frame rates (e.g. 60 fps) within the bandwidth of the fluoroscope's electronics. When this occurs a step in spatial resolution is expected when the frame rate is reduced.

Because noise perceptibility is increased with increasing magnification projected onto the observer's eye, it is necessary to increase radiation dose as magnification is increased. Essentially, visual noise limitation requires that the same number of photons are used to form a pixel irrespective of its size. As discussed above, image intensifier systems require an increase in radiation dose rate proportional to the ratio of the square of the FOV with image-intensifier magnification to account for the physical loss of minification gain when a smaller FOV is selected. Ignoring any bundling-unbundling transition, changing the zoom on a FPS with a corresponding secondary magnification by the system's image processor is usually accompanied by a deliberately programmed increase in patient dose rate. It was empirically found that increasing fluoroscopic dose rate inversely proportional to the linear size of the active FOV produced satisfactory clinical results for both fluoroscopy and acquisition modes.

XV. DIGITAL IMAGE PROCESSING (2010-2019)

Fluoroscopic X-Ray tubes and image receptors have not changed in any fundamental manner in the past decade. Increased image processing power has had a substantial impact on both clinical conspicuity and radiation use in the same time interval.

The job of the image acquisition elements of the fluoroscopic system is to deliver technically adequate images to the system's digital image processor. These images are then processed, and the results delivered to the observer. In modern terminology, these image stages are called 'for processing' and 'for presentation'. As discussed below, presentation images usually undergo non-linear transformations, that are configured based on presumed clinical content. Many fluoroscopes delivered after 2010 have enough computational resources to provide different algorithms in different regions of the same image. These algorithms rely on a-priori knowledge of the examination in progress. Such information is currently supplied by the operator's selection of an examination-set. A future fluoroscope might improve selections by analysis of the patient's current and historical images.

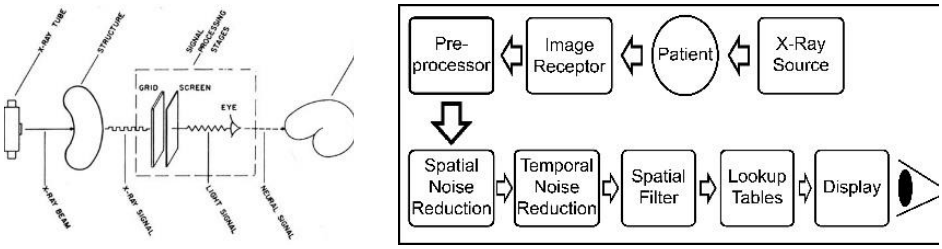
Figure 76 sketches information flow in 1966 and 2016 fluoroscopes. In both eras, X-rays interact with a patient to produce a modulated beam which is then detected and converted into a visible image projected into the observer's eye-brain system. The characteristics of the 1966 image were fixed by a combination of imaging hardware and operating conditions (e.g. kVp, mAs). In 2016, the X-ray source has additional controls that affect images (e.g. variable framerate, and shaped X-ray

spectrum). Also, controllable image processors and displays are in the information flow line between the imaging system and the observer.

Fig 57: Fluoroscopic Information Flow

a) Morgan (1966) – permission pending

b) based on Lendil (2018)



Global temporal averaging is an inherent characteristic of analog video tubes. As discussed above, changing tube types altered both noise characteristics and motion blur. The lack of lag in newer camera types increased the appearance of noise, including X-ray quantum noise. Image processors can supply temporal averaging using the recursive filtering algorithm. Figure 58 illustrates the effects of adding recursive filtering. The original images were acquired using pulsed fluoroscopy with short pulses. There is minimum motion blur of the rapidly moving guidewire in the original images (it is in a different position in each image). The number of older images contributing to the current image increases with increased recursion. This results in ghosting of previous guidewire positions (arrow).

Fig 58: Global Temporal Recursive Filtering

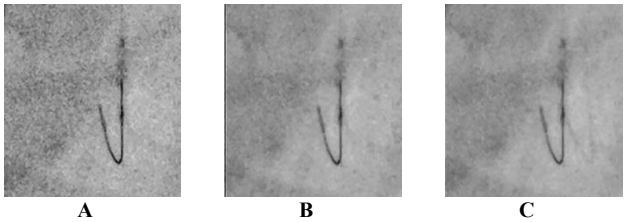
a) No recursion

b) Low recursion

c) High recursion

Visual noise decreases and ghosts of moving objects increase with increased recursion

Source- Siemens Healthineers - Lendil



Many algorithms can be used to modify the global appearance of the images for the purpose of increasing the perceptibility of clinically important information. Four of these are sketched in Figure 59.

Fig 59: Image processing algorithm components

a) Single frame smoothing

In this example, a simple gaussian blur has been used to process the right-hand image shown in Fig 58

b) Temporal (Multi frame) smoothing discussed in text above

c) Automatic motion compensation

d) Multi-parameter image enhancement

b-d) Source – Philips Healthcare - Jans

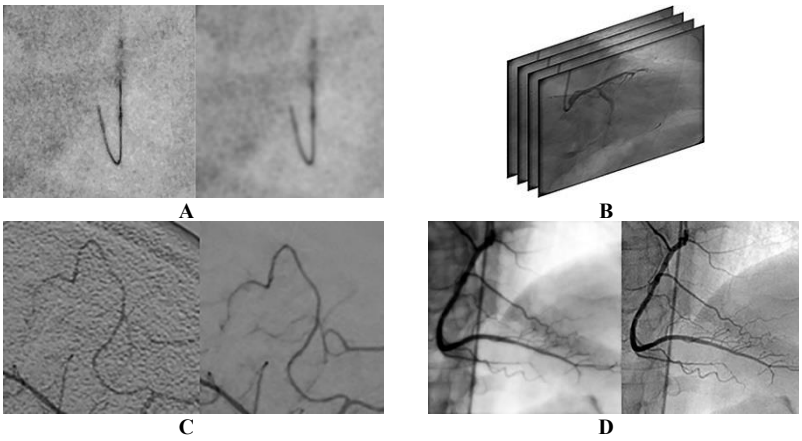
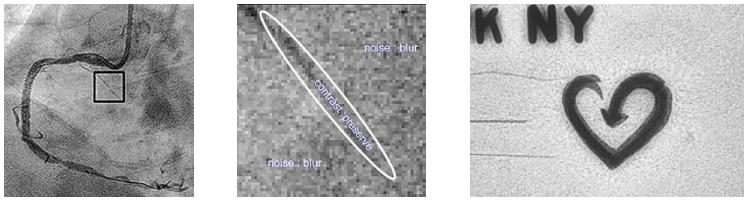


Image processors are now fast enough to apply different algorithms in different parts of the same image, and in different but corresponding parts of an image sequence. Figure 60 illustrates single-frame local processing. The image processor has identified the small contrast filled artery in the image on the left. On the right, the original pixel data is unchanged near the vessel and pixel averaging is applied to smooth the background appearance.

Fig 60: 2018 Single Frame Local processing

a,b) Clinical Image. Less spatial averaging is applied near the artery. Source: Philips Healthcare - Jans

c) Last-image-hold of a portion of a daily fluoroscopic QA phantom. Less spatial averaging



near lead markers and guide wires, more in uniform areas of the image. Source – Columbia

A **B** **C**

Figure 61 is a photographic simulation of multi-frame local processing. The local focus in this simulation is a vertical section of the right coronary artery. Noise was separately added to three replicas of the same initial image before combining them.`

Fig 61: Multi- Frame (temporal) Local Processing Simulation

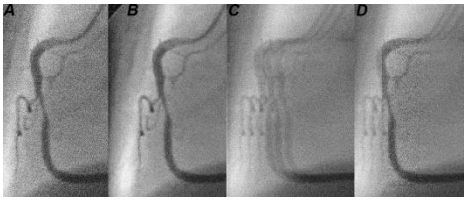
a) Single image

b) Three replicas summed without motion

c) Replicas summed with simulated motion.

d) Replicas summed except for the region near the vertical RCA segment, single image data used here.

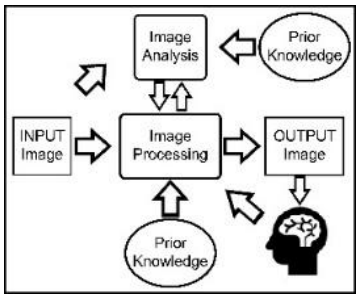
Note the ghosting in summed areas in c and d; note the increased noise near the RCA in d



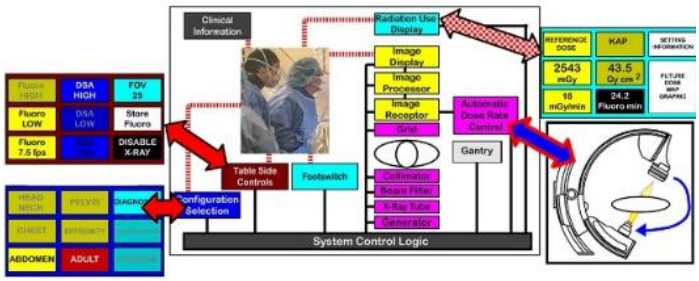
The importance of prior and real-time control of fluoroscopic image processing will continue to grow. The operator is a key node in managing automatic image analysis and processing as well as in other aspects of controlling and optimizing the settings and use of the entire fluoroscopic system. (Fig.62). The starting point is the pre-procedure transfer of patient current and historical data, procedural intent, operator preference, and other factors.

Conditions can and will change during a procedure. The goal is to continuously optimize both image acquisition and image processing parameters when these changes occur. Some (e.g. substituting CO_2 for Iodine as a contrast medium) should be detectable by automatic image analysis. Others (e.g. replacing a steel angiographic guidewire with a Platinum plated version) may not be recognizable by the image processor. The challenge is to maintain optimization without unnecessarily distracting operators and their assistants from essential patient care.

Fig 62: A-priori, automated, and operator- controlled image processing



a) Image processing and analysis uses both prior knowledge and feedback from the observer – Based on Lendil 2018



b) The operator is a key node in many fluoro control loops.

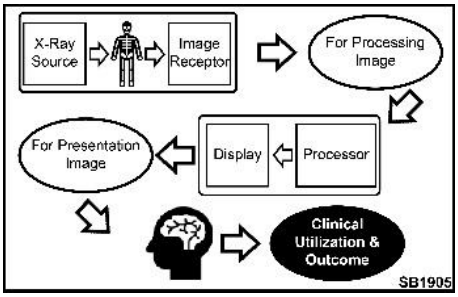
XVI. COMMENT

Radiologic imaging procedures needed to achieve a clinical goal should be performed in a manner that optimizes the balance between potential clinical benefits and all risks (not just minimize radiation)[58]. Fluoroscopic optimization differs from most other modalities in that achieving patient benefits imposes risks on both patients and staff.

Information flow can broadly be segmented into acquisition, processing, display, and utilization (Figure 63). Each of these domains should be optimized by itself and with consideration of the other domains. Equipment quality testing has traditionally focused on acquisition. The availability of 'for-processing' images between the acquisition and processing domains provides the opportunity to develop better testing tools for the imaging hardware and its configuration. Potentially, a digital test image (SMPTE is one simple example) could be inserted into the image processor and separately test the processing-display portion of the system. Metrics describing the operators' use of the fluoroscope and the clinical result of such use are under development[58]. Image quality evaluations might eventually include short-term and long-term clinical outcomes from large numbers of procedures.

Fig 63. Information Flow

The for-processing, for-display, and clinical-utilization nodes could be used for quality evaluations.



Fluoroscopic technology has progressed over the past 125 years. In the first decade, the operator needed a considerable body of physical and technical knowledge in order to adjust the system to for it to work at all. In the most recent decade, the operator's main challenge is to supply the system with necessary and sufficient information so that a well configured fluoroscope can self-optimize to meet immediate clinical needs. A consequence of this requirement is that equipment designers, service personnel, applications specialists, and medical physicists must be aware of the specific imaging requirements needed to meet different clinical needs. Achieving and maintaining an appropriate level of clinical understanding is one of the major tasks for continuously improving imaging in the 21st century.

XVII. REFERENCES

1. W. C. Roentgen, Aus den Sitzungsberichten der Würzburger Physik.-medic. Gesellschaft Würzburg, 137-147 (1895).
2. W. C. Roentgen, Aus den Sitzungsberichten der Würzburger Physik.-medic. Gesellschaft Würzburg, , 11-17 (1896).
3. W. C. Roentgen, Mathematische und Naturwissenschaftliche Mitteilungen aus den Sitzungsberichten der Königlich Preußischen Akademie der Wissenschaften zu Berlin, 392-406 (1897).
4. ANON, Nature 54 (1388), 109-112 (1896).
5. S.J.R., Nature 54 (1409), 621-621 (1896).
6. ANON, Scientific American (April 4, 1896), 219 (1896).
7. G. King, (smithsonian.com, 2012), Vol. 2019.
8. ANON, Scientific American (August 8, 1897), 88-89 (1897).
9. T. C. Gilcrist, Bull Johns Hopkins Hosp 8 (71) (1897).
10. S. Tousey, MEDICAL ELECTRICITY AND RONTGEN RAYS WITH CHAPTERS ON PHOTOTHERAPY AND RADIUM, first ed. (W. B. SAUNDERS COMPANY, PHILADELPHIA, 1910).
11. I. Rock, An Introduction to Perception. (Macmillin Publishing Company, New York, 1975).
12. B. A. Wandell, Foundations of Vision. (Sinauer Associates Inc, Sunderland, MA, 1995).
13. W. E. Chamberlain, Radiology 38 (4), 383 - 413 (1942).
14. F. J. Hodges, Radiology 38 (4), 453-461 (1942).
15. R. E. Sturm and R. H. Morgan, American Journal of Radiology 62 (1949).
16. A. Rose, J Opt Soc Am 38 (2), 196-208 (1948).
17. J. W. Coltman, Radiology 51 (3), 359-367 (1948).
18. NBS, 1963.
19. T. Holm and R. D. Moseley, Jr., Radiology 82 (5), 898-904 (1964).
20. R. H. Morgan, Radiology 86 (3), 403-416 (1966).
21. N. R. Silverman, Radiology 103 (2), 263-265 (1972).
22. B. M. Lantz, J. M. Foerster, D. P. Link and J. W. Holcroft, AJR Am J Roentgenol 134 (6), 1161-1168 (1980).
23. D. M. Hynes, E. W. Edmonds, K. R. Krametz and D. Baranoski, Radiology 133 (3 Pt 1), 751-755 (1979).
24. S. Balter, F. M. Sones, Jr. and R. Brancato, Circulation 58 (5), 925-932 (1978).
25. E. W. Gertz, J. A. Wisneski, R. G. Gould and J. R. Akin, Am J Cardiol 50 (6), 1283-1286 (1982).
26. H. Dash and D. M. Leaman, J Am Coll Cardiol 4 (4), 725-728 (1984).
27. M. R. Pitney, R. M. Allan, R. W. Giles, D. McLean, M. McCredie, T. Randell and W. F. Walsh, J Am Coll Cardiol 24 (7), 1660-1663 (1994).
28. L. E. Watson, M. W. Riggs and P. D. Bourland, Health Phys 73 (4), 690-693 (1997).
29. S. Balter, Radiat Prot Dosimetry 94 (1-2), 183-188 (2001).
30. E. Kuon, M. Schmitt and J. B. Dahm, Am J Cardiol 89 (1), 44-49 (2002).
31. E. Kuon, M. Gunther, O. Gefeller and J. B. Dahm, Rofo 175 (11), 1545-1550 (2003).
32. A. Komemushi, N. Tanigawa, S. Kariya, H. Kojima, Y. Shomura and S. Sawada, J Vasc Interv Radiol 16 (10), 1327-1332 (2005).
33. N. T. Fitousi, E. P. Efstathopoulos, H. B. Delis, S. Kottou, A. D. Kelekis and G. S. Panayiotakis, Spine (Phila Pa 1976) 31 (23), E884-889; discussioin E890 (2006).
34. B. A. Schueler, T. J. Vrieze, H. Bjarnason and A. W. Stanson, Radiographics 26 (5), 1533-1541; discussion 1541 (2006).
35. O. Dragusin, R. Weerasooriya, P. Jais, M. Hocini, J. Ector, Y. Takahashi, M. Haissaguerre, H. Bosmans and H. Heidbuchel, Eur Heart J 28 (2), 183-189 (2007).
36. K. P. Kim and D. L. Miller, Radiat Prot Dosimetry 133 (4), 227-233 (2009).
37. L. W. Klein, D. L. Miller, S. Balter, W. Laskey, D. Haines, A. Norbash, M. A. Mauro and J. A. Goldstein, Radiology 250 (2), 538-544 (2009).
38. B. A. Schueler, Tech Vasc Interv Radiol 13 (3), 167-171 (2010).
39. O. P. Haqqani, P. K. Agarwal, N. M. Halin and M. D. Iafrati, J Vasc Surg 55 (3), 799-805 (2012).
40. D. L. Miller, Health Phys 105 (5), 435-444 (2013).
41. G. Christopoulos, A. C. Papayannis, M. Alomar, A. Kotsia, T. T. Michael, B. V. Rangan, M. Roesle, D. Shorrock, L. Makke, R. Layne, R. Grabarkewitz, D. Haagen, S. Maragkoudakis, A. Mohammad, K. Sarode, D. J. Cipher, C. E. Chambers, S. Banerjee and E. S. Brilakis, Circ Cardiovasc Interv 7 (6), 744-750 (2014).

42. A. den Boer, P. J. de Feyter, W. A. Hummel, D. Keane and J. R. Roelandt, Circulation 89 (6), 2710-2714 (1994).
43. R. M. Gagne, P. W. Quinn and R. J. Jennings, Med Phys 21 (1), 107-121 (1994).
44. M. P. Capp, Radiology 138 (3), 541-550 (1981).
45. D. L. Ergun, C. A. Mistretta, R. A. Kruger, S. J. Riederer, C. G. Shaw and D. P. Carbone, Radiology 132 (3), 739-742 (1979).
46. R. A. Kruger, C. A. Mistretta, T. L. Houk, W. Kubal, S. J. Riederer, D. L. Ergun, C. G. Shaw, J. C. Lancaster and G. G. Rowe, Invest Radiol 14 (4), 279-287 (1979).
47. A. B. Crummy, C. M. Strother, J. F. Sackett, D. L. Ergun, C. G. Shaw, R. A. Kruger, C. A. Mistretta, W. D. Turnipseed, R. P. Lieberman, P. D. Myerowitz and F. F. Ruzicka, AJR Am J Roentgenol 135 (6), 1131-1140 (1980).
48. C. G. Shaw, D. L. Ergun, R. A. Kruger, C. A. Mistretta, A. B. Crummy, D. Myerowitz, C. M. Strother, J. Sackett, M. Van Lysel, W. Zarnstorff and W. Turnipseed, Nuclear Science, IEEE Transactions on 27 (3), 1042-1046 (1980).
49. S. Balter, D. Ergun, E. Tscholl, F. Buchmann and L. Verhoeven, Radiology 152, 195-198 (1984).
50. D. S. Groth, S. N. Bernatz, K. A. Fetterly and N. J. Hangiandreou, Radiographics 21 (3), 719-732 (2001).
51. G. Pang, W. Zhao and J. A. Rowlands, Med Phys 25 (9), 1636-1646 (1998).
52. J. H. Siewerdsen, L. E. Antonuk, Y. el-Mohri, J. Yorkston, W. Huang and I. A. Cunningham, Med Phys 25 (5), 614-628 (1998).
53. M. Strotzer, J. Gmeinwieser, M. Volk, R. Frund, J. Seitz, C. Manke, H. Albrich and S. Feuerbach, AJR Am J Roentgenol 171 (1), 23-27 (1998).
54. M. Strotzer, M. Volk and S. Feuerbach, Electromedica 66 (2), 52-57 (1998).
55. H. D. Kubo, E. G. Shapiro and E. J. Seppi, Med Phys 26 (11), 2410-2414 (1999).
56. N. Matsuura, W. Zhao, Z. Huang and J. A. Rowlands, Med Phys 26 (5), 672-681 (1999).
57. S. Balter, Catheter Cardiovasc Interv 63 (3), 331 (2004).
58. S. Balter, M. Brinkman, S. Kalra, T. Nazif, M. Parikh, A. J. Kirtane, J. Moses, M. Leon, A. Feri, P. Green, Z. A. Ali, M. Liao and D. Karmpaliotis, EuroIntervention 13 (12), e1468-e1474 (2017).

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THE SCIENTIFIC AND TECHNOLOGICAL DEVELOPMENTS IN MAMMOGRAPHY
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Abstract— **The human breast, unlike most other anatomical regions of the body, is composed of normal and pathologic tissues with very small differences in physical density as a source of contrast for x-ray imaging. This low physical contrast cannot be adequately imaged with conventional radiographic methods. Small calcifications, a physical sign associated with some cancers, are also not visible because of the normal blurring within the radiographic process. Effective mammography, *breast radiography*, requires an imaging procedure with high contrast sensitivity to visualize the soft tissue structures and low blurring to enhance visibility of calcifications. The development of this capability has been an ongoing effort for over a half century. Along with developments for increased visibility within the breast there has been progress in optimizing image quality with respect to radiation dose to patients and improving the efficiency of the total mammography process. Transitioning from conventional radiography to mammography required innovations and developments in two specific areas, the x-ray beam *spectrum* and the imaging *receptor*. An x-ray spectrum that was more optimum for mammography used specific anode materials, molybdenum and rhodium and filters of the same elements. The two major requirements for the image receptor that required years of ongoing development were a wide latitude/dynamic range to capture and display contrast, and very low blurring to provide visibility of the small calcifications. After the initial development of mammography using industrial radiographic film exposed directly with the x-radiation, intensifying screens specific for mammography were developed. This was along with developments of film and film processing to be used with the intensifying screens. This development transitioned through several phases including the transition from calcium tungstate to rare earth screen materials and film requiring viewing conditions different from conventional radiography. X-ray tubes with small focal spots were in the dedicated systems developed specifically for mammography. This included a very small focal spot used with geometric magnification to decrease the effective receptor blur and provide the highest visibility of detail of any medical imaging procedure to enhance the visibility of the small calcifications. The approaches to and methods for re-shaping or compressing the breast during the imaging procedure evolved over time. In general, the first half-century of developments in mammography, the subject of this article, used film as a receptor component, archive medium, and display. The development of digital imaging technology provides solutions to some of the challenges in mammography technology and procedures and brings that phase of mammography technology to a conclusion.**

Keywords— **Mammography, breast cancer, image quality, x-ray spectrum, film.**

CONTENT

I. INTRODUCTION

Mammography**,** radiography of the breast, is a major medical procedure for detecting, diagnosing, and managing the treatment of cancer and other breast pathologies. What might appear to be a relatively simple x-ray procedure is in reality a complex process that has faced many challenges in its development over the years. Mammography has a significant and interesting history that has been well researched and documented by others. These authors have focused on diverse topics that define the continuing development of mammography including: clinical applications, professional practice and accreditation, extensive quality assurance activities, political and regulatory requirements, issues in public media, and the highly-significant contributions in the fields of *physics* and developments in *technology*. Our specific interest here is on the physics and technology. Much of this history has been published and is included in the bibliography and addendum at the conclusion of this article and will not be duplicated here. Those articles on the history provide extensive references to publications reporting the research and developments that have resulted in the continuing evolution of mammography. In this article our focus is more on the "why" rather than the just "what" was done in the continuing development of mammography, looking from a physics and technology perspective. This is consistent with the experience of the author, who as a physicist has been involved in the development and clinical applications of mammography for most of its active history.

This focus can almost be summarized in one phrase, the *quest for visibility*. The purpose of a mammography procedure is to provide physicians with the ability to see, or visualize, the internal anatomical structures and potential signs of pathology, especially cancer, within a breast. Here is the challenge. The breast, unlike most of the other regions of human body, is composed of soft tissue with very small differences in physical density that are the source of contrast for imaging. Also, a significant signs of some breast cancers are very small "micro" calcifications that are beyond the visibility of detail capability of most medical imaging methods.

To meet these challenges an effective mammographic procedure must have high *contrast sensitivity* to visualize the soft tissues, including cancers, and extremely *low blurring* for imaging the small calcifications. There is also the goal of minimizing radiation exposure and dose to the patient but with the recognition of the conflicting relationship between image quality and radiation exposure.

A major advancement was the transition from using conventional x-ray equipment to the development of *dedicated* machines specific for mammography.

It was these requirements that defined the continuing development of mammography technology and physics applications that we will now explore.

We begin with an overview of the mammography process in Figure 1.

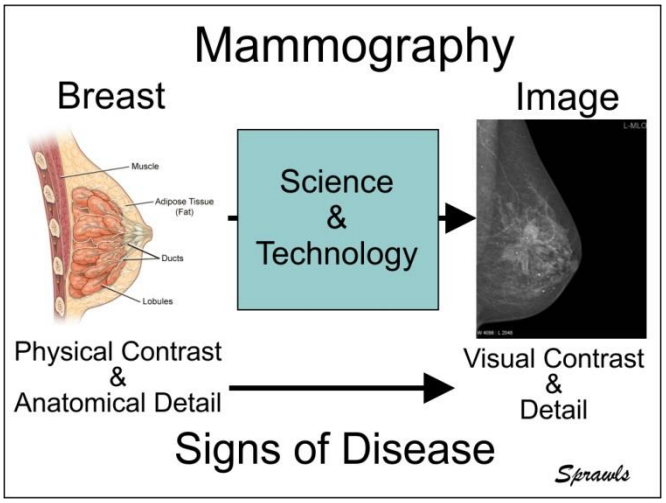


Figure 1. Mammography is based on the application of science, especially physics, and continuing developments in technology, to provide high-quality visualization of the interior of the human breast anatomical structures and signs of disease within a breast.

II. THE EVOLVING ELEMENTS OF MAMMOGRAPHY PHYSICS AND TECHNOLOGY

Mammography is a radiographic procedure. Radiography has been used for imaging most anatomical regions of the human body following Roentgen's discovery and extensive research in 1875. Most of the body, and especially with the

introduction of barium and iodine contrast media, could be imaged with x-radiation. The breast was an exception. There were efforts to do mammography with the available radiographic methods but with limited clinical results. What were needed were modifications of virtually every component of the radiographic system to enable imaging with both high contrast sensitivity and very low blurring. Figure 2 shows the elements of a mammographic system that have been researched and developed in the continuing evolution of mammography.

The two major elements in mammography that are very different compared to general radiography for all other parts of the body, are the *x-ray beam spectrum* and the *receptor and image display*. It is the continuing research and development to improve and optimize these two components of the mammography system that form much of the history that we will now explore.

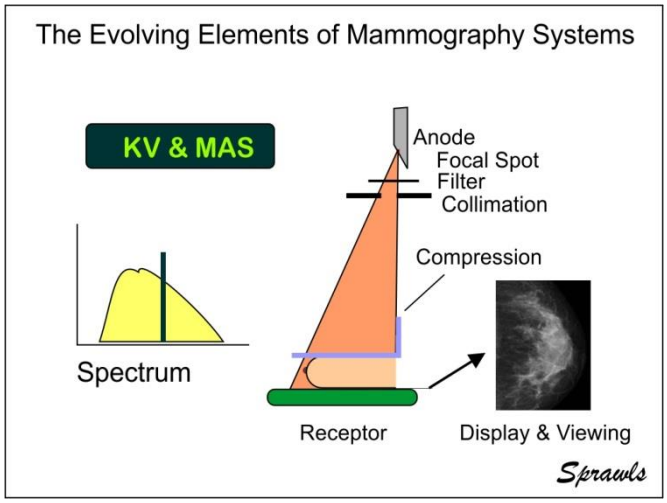


Figure 2. The elements of the mammography system that have evolved over the years in the effort to provide better visualization within the breast.

The evolution of mammography into a highly-effective method for diagnosing and reducing deaths from breast cancer is not just the development of equipment. It was a collaborative effort including extensive clinical research, development of methods and procedures, demonstration of its effectiveness and value, promotion within the medical profession, and major educational efforts. It is these combined efforts by Robert Egan, M.D. that resulted in his recognition as "The Father of Mammography" and provided the foundation for the continuing developments in mammography.

III. THE EGAN METHOD

Dr. Egan joined the faculty of Emory University in Atlanta in 1965, coming from the M.D. Anderson Cancer Center in Houston where he began his pioneering work in mammography while he was in his residency training. It was at Emory where the author of this article, along with several other physicists, were his collaborators in research, continuing development, and clinical applications of innovations in mammography. The so-called Egan Method included the development of an x-ray imaging system specific for mammography along with the development of imaging procedures and techniques. These, along with extensive descriptions of the clinical characteristics and diagnosis of breast cancer, were described in textbooks he authored both for physicians and technologists shown in Figure 3.

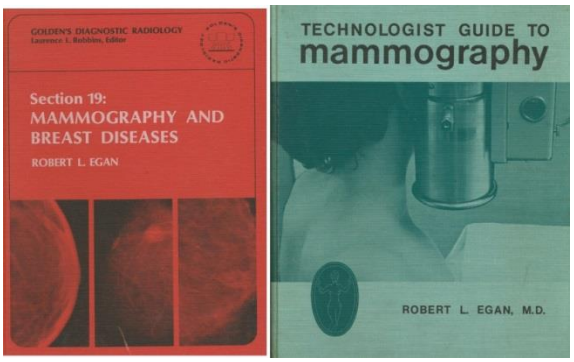


Figure 3. Two of the books by Egan describing the clinical and technical procedures in mammography.

It was these books along with courses by Egan that were a major factor in establishing mammography as a major and valuable medical procedure.

We will now consider the imaging technology and process developed by Egan and collaborating physicists and use it as a reference for the continuing evolution and advances of mammography.

The mammography system can be considered as three major elements that have evolved over the years in the quest of increased visibility and image quality. These are:

- Geometry, Spatial Relationships, and Configuration of the Breast
- The X-Ray Spectrum, Contrast Sensitivity and Radiation Dose
- Image Receptor, Processing, and Viewing

IV. GEOMETRY, SPATIAL RELATIONSHIPS, AND CONFIGURATION OF THE BREAST

This has been one of the major factors that have evolved in the continuing development of mammography technology, especially with the move from the use of conventional radiography to *dedicated* mammography systems. The development of dedicated mammography systems addressed all of the elements listed above with the goal of improving image quality and capability for positioning and obtaining views in clinical procedures.

In The Beginning

The geometry and positing in the Egan method is shown in Figure 4.

Figure 4. The positioning developed by Egan provided three anatomical views considered necessary to properly visualize features and potential cancers within the breast as shown here in his textbook.

The Egan method illustrated here was using modified conventional x-ray equipment. The relatively long x-ray tube to receptor distance reduced focal spot blurring to enhance the visibility of calcifications. What is prominent here, especially from a more recent perspective, is there is no compression of the breast. That develops later.

The Evolution of Breast Compression

The natural shape of the non-confined breast as illustrated in Figure 5 presents a challenge to maximum image quality in several ways.

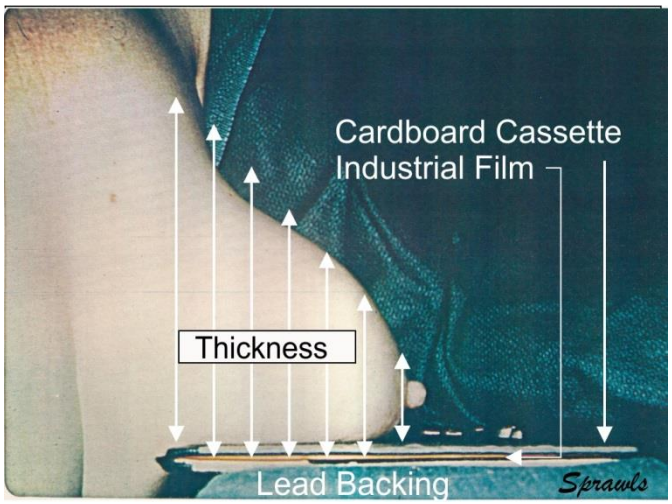


Figure 5. An early mammography receptor consisting of industrial type radiographic film in a cardboard cassette with a lead backing to reduce scattered radiation from the support. Also illustrating the wide range of breast thickness to be imaged in this photograph by the author.

The variation in thickness from the chest wall to the nipple creates a wide range of exposures to the receptor that can extend beyond the latitude/dynamic range of film and result in reduced contrast as will be discussed later.

Adequate visualization of micro calcifications requires the total blurring to be limited to approximately 0.15 mm. Exposure times in mammography can be several seconds, very long compared to most other radiographic procedures. Almost any patient motion during the exposure can be detrimental to image quality.

Physical compression and stabilization of the breast was developed to improve image quality both with respect to contrast and reduced blurring. The general evolution of compression is illustrated in Figure 6.

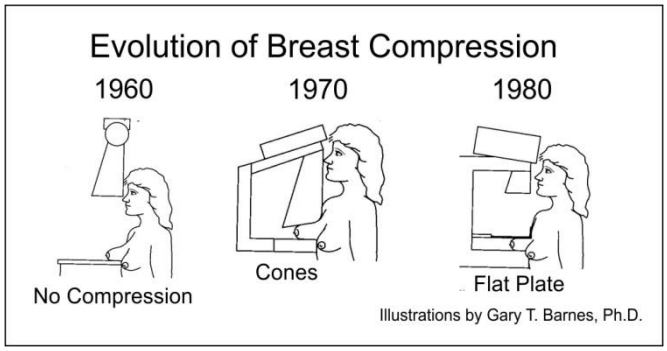


Figure 6. The three major phases of imaging geometry and breast compression.

A continuing challenge to breast compression, especially very firm compression, is discomfort to the patient. In the 1960s, when conventional x-ray tubes with tungsten anodes were used, the x-radiation was generally more penetrating than with the later molybdenum anode and filters. When using the directly exposed film developed for industrial radiography the latitude or dynamic range issue appeared to be less of a problem, so compression was not considered necessary for adequate quality.

After that, and as x-ray systems were developed specific for mammography, some type of compression and breast stabilization was included by the x-ray beam cone and sometimes with soft components in contact with the breast.

The introduction and continuing development of flat plate compression illustrated in Figure 7 was a major contribution to image quality in several ways.

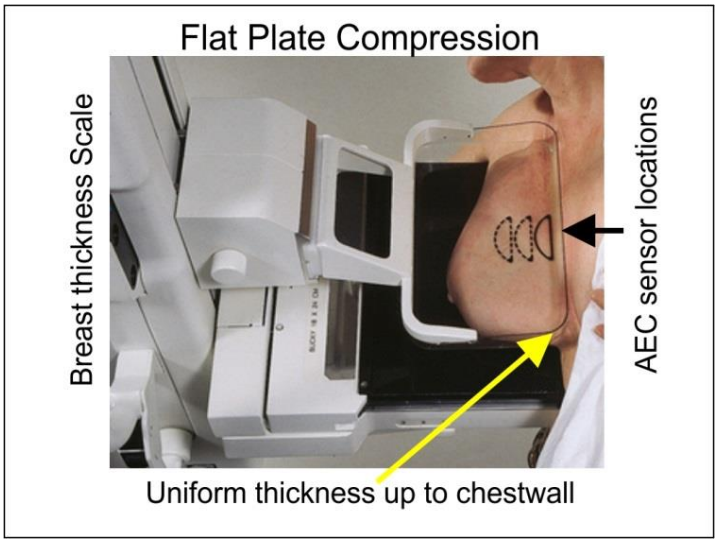


Figure 7. Features associated with flat plate breast compression devices that contribute to improved image quality are identified in this illustration provided by the author.

The stabilization of the breast that eliminated most patient motion reduced that potential source of blurring and enhanced visibility of calcifications. However, the major contribution is the re-shaping of the breast into a more uniform thickness spread over a slightly larger area. The uniform thickness reduced the range of x-ray exposure to the receptor and loss of contrast because of film latitude limitations. The spreading of the breast tissue, especially in the thicker regions, reduced the overlapping of objects and structures that could potentially interfere with visualization. This was especially significant for imaging the tissues up to the chest wall.

There are several other features in the development of the compression system that aided the technologist in producing high-quality images. One was the measurement and display of the compressed breast thickness that was a major factor in selecting optimum technique factors. Also, the diagram on the compression plate guided the selection of the appropriate automatic exposure control (AEC) sensor location for each specific patient procedure. The sensors were located below the receptor so that they did not interfere with the image. By selecting which sensor location to use the technologist determined the area in the breast that would result in the desired film density.

The evolution of breast compression is a significant element in the history of mammography. It has focused on modifying the anatomical environment of the breast for optimum imaging. In spite of its contribution to high-quality imaging, the compression technique is a source of discomfort and pain to patients. The challenge continues.

V. THE X-RAY SPECTRUM, CONTRAST SENSITIVITY AND RADIATION DOSE

The production of a visible image depends on the ability to "see" physical contrast within the body and convert that into visible contrast within an image. Physical contrast is the difference in *physical density* among the tissues and body structures with some contribution from differences in atomic number (Z). The bones within the body were the first to be imaged (beginning with Mrs. Roentgen's hand) because of the high physical contrast between the calcium in the bones and the soft tissues in the body. It was soon discovered that the chest could be imaged because the low-density air within the lungs provided an excellent background for the more dense bones, fluid, and signs of disease within the lungs. With the development of contrast agents containing barium and iodine with their desirable atomic numbers for x-ray attenuation, the scope of imaging, both radiography and fluoroscopy, was expanded to include virtually all regions and systems of the body, *except for the breast.*

The Challenge of Breast Imaging

X-ray imaging of the breast—mammography—faced many challenges and required many years of research and development to reach its full potential. The major factor is that the breast is composed of soft tissues with small differences in density and physical contrast both among the normal anatomical structures and abnormal tissues, especially cancers. The visualization of these requires a procedure with higher *contrast sensitivity* than more conventional radiographic procedures.

The physics of x-ray image formation was well established with the known dependency of x-ray attenuation, specifically the photoelectric effect, and the formation of contrast among low atomic number soft tissues inversely related to photon energies. An x-ray spectrum with low photon energies would be required to produce adequate contrast and visibility among the soft tissues, both normal and pathologic tissues including cancer. There was also the factor that the lower photon energies were less penetrating through the total breast and resulted in increased exposure and dose to the breast. Both of these factors, image contrast and dose, also depended on the thickness and density of the breast. This was to be a major challenge to be addressed throughout much of the development of mammography with two questions:

- What is the optimum spectrum for imaging a specific breast size?
- How to produce an x-ray beam with that spectrum?

The Optimum X-ray Spectrum

That question began to be answered by Gajewski, H & H Reiss, K. Physical fundamentals and technique in soft tissue diagnosis. Der Radiologe. 14. 438-46. (1974). With their innovative research and results shown in Figure 8.

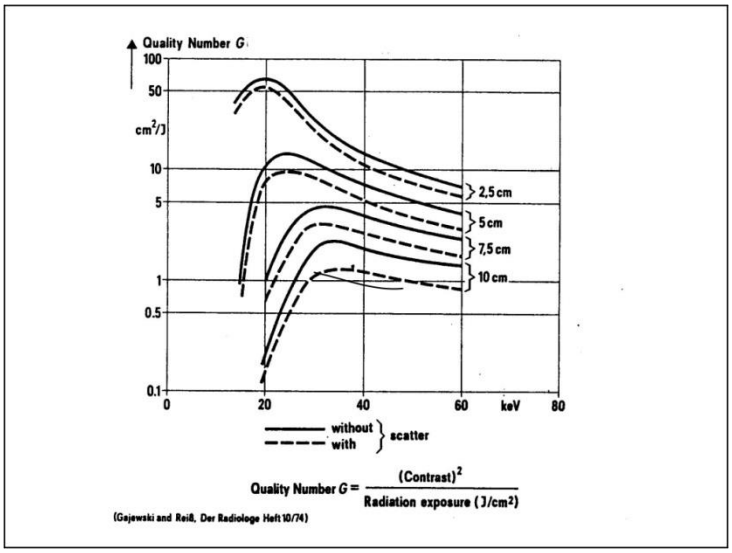


Figure 8. A quality number representing the ratio of contrast to dose displayed over a range of photon energies.

These experiments conducted with varying thicknesses of water simulating a range of breast sizes with measurements over a spectrum up to 60 keV demonstrated several factors that were to guide the developments in mammography for many years to come. Perhaps the most significant was optimum photon energy for specific breast sizes and that it increased with size. Also, as breast size increased, the ratio of quality to exposure decreased, with scattered radiation becoming a more significant factor.

This demonstrated that for the expected range of breast sizes, especially when compressed, of 2.5 mm to 7.5mm, the optimum photon energy was in the range of 20 keV to 30 keV. The challenge was how to produce x-ray spectra to fulfill these requirements. A reasonable assumption is that an x-ray machine that produces a mono-energy spectrum that could be adjusted with respect to breast thickness and density would be the "ideal" system. However, that is yet to be developed.

In principle, the x-ray spectrum should be adjusted to be *optimum* for each patient in relation to breast thickness and density. For many years the anode and filter materials were determined by the design and construction of the equipment and could not be changed by the operator, leaving KV as the adjustable technique factor.

Tungsten and Minimum Filtration

In the beginning x-ray tubes with tungsten anodes were available and used for all medical imaging procedures. The imaging system developed by Egan and his physics team consisted of conventional tungsten anode x-ray tubes with the added filter removed leaving only inherent tube window material.

X-ray tubes with beryllium windows were becoming available and used in mammography. Beryllium has an atomic number of 4 and a relatively low density (1.85 g/cm3) minimizing its x-ray attenuation, especially for the lower-energy photons. In some applications beryllium window tubes were used without additional filtration but with concern for high exposures to the breast. The ultimate advantage of beryllium window tubes was permitting other types of filters to be added that were more appropriate for breast imaging, as described later.

At this time mammography was performed with film as the receptor exposed directly with x-radiation without intensifying screens to minimize blur. This required a relatively high x-ray exposure. In addition to producing radiation over a long exposure time, up to 6 seconds, a major requirement for the modified equipment was the capability of providing low, adjustable, and accurate KV values over the range of 22 kV to 34 k V.

In addition to modifying the x-ray tubes with respect to filtration the generators or power supplies require some changes in design. For effective mammography they were required to operate at lower KV values than for conventional radiography and with good accuracy and produce high tube currents, MA, over a relatively long exposure time. Initial experiments have used modification of some existing X-ray equipment used for radiography (50 to 120 kV), grounding one side of the high voltage generator, thus producing half of the kV range (25 to 60 kV), using the existing kV regulation of the generator.

Typical Egan technique factors for a medium size breast was 28 kV, 300 mA, 6 seconds (1800 mAs), and a FRD of 36 in.

Figure 9 shows physicists in Egan's laboratory at Emory University analyzing the performance of the x-ray generators being developed for mammography.

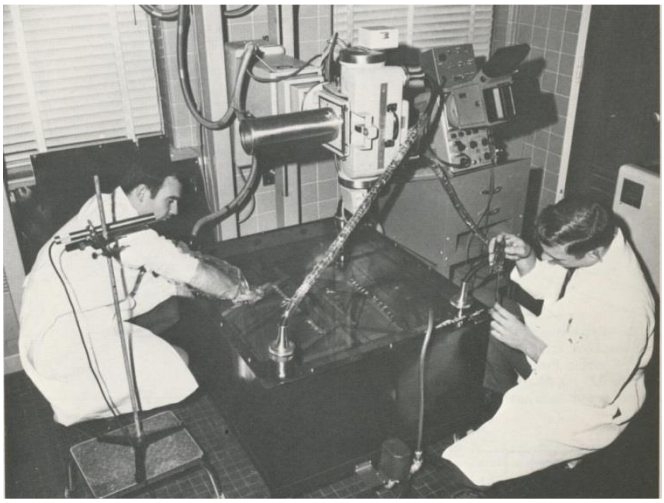


Figure 9. Physicists using a bank of resistors in a tank of insulating oil as an electrical load to evaluate the performance of x-ray generators being developed for mammography

The Significance of KV on Procedure Optimization: Image Quality and Radiation Dose

Conventional radiography equipment, especially with modifications as described, played a role in establishing mammography as a valuable medical procedure. However, limitations were realized. One of these was the necessity to have accurate and precisely controlled KV values in the general range from 24 kV to 34 kV. The KV value was to become the major adjustable technique factor by the technologists in relationship to the thickness and density of individual patient breasts. Differences in KV values as little as 2 kV were significant in optimizing a procedure with respect to quality and dose.

The inclusion of generators/power supplies that could meet these KV requirements was one of the major features of the *dedicated* mammography systems to be developed. Also, measuring and evaluating KV accuracy became a required quality assurance function performed by medical physicists.

The Impact of Molybdenum: Anode and Filters

Molybdenum is a metal with a high melting point and an atomic number (Z) of 42. It is the combination of these two characteristics, especially the atomic number that has made molybdenum a major element in mammography both as an x-ray tube anode material and x-ray beam filter. When used together the x-ray spectrum shown in Figure 10 is produced.

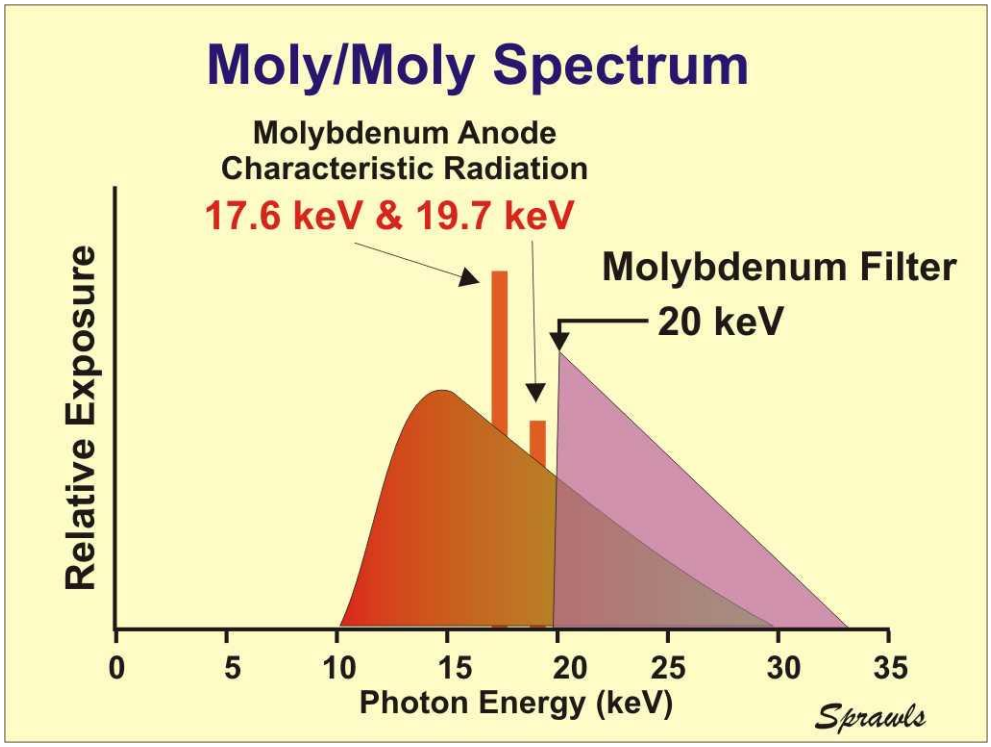


Figure 10. The x-ray spectrum produced with the combination of a molybdenum anode and filter.

The significant characteristic of molybdenum with its atomic number (Z) of 42 is that the anode produces characteristic x-ray peaks at 17.6 keV and 19.7 keV and a filter attenuation K edge at 20 keV as shown. It was with this introduction that *characteristic radiation*, and not bremsstrahlung, became a major component of mammography.

It is the combination of the molybdenum anode and filter that produces an x-ray spectrum within a relatively narrow range of energies near 20 keV that makes it optimum with respect to image contrast and radiation dose to patients, especially for smaller breasts as indicated in Figure 8.

The molybdenum anode and filter was, and continues, to be the foundation of x-ray breast imaging. Some developments described later shifted the spectrum to slightly higher energies that were more optimized for larger and denser breasts.

Moving Up to Rhodium

It was recognized that the molybdenum – molybdenum (anode and filter) aka "moly-moly" spectrum, while appropriate for smaller breasts, was not optimum for all. A solution was provided by the element rhodium. It has some of the same metallic properties as molybdenum, including a high melting point. However, its atomic number (Z) of 45, compared to 42 for molybdenum, shifts both the characteristic radiation and the K-edge energies up to higher values. This is optimum for larger and denser breasts. This spectrum is shown in Figure 11.

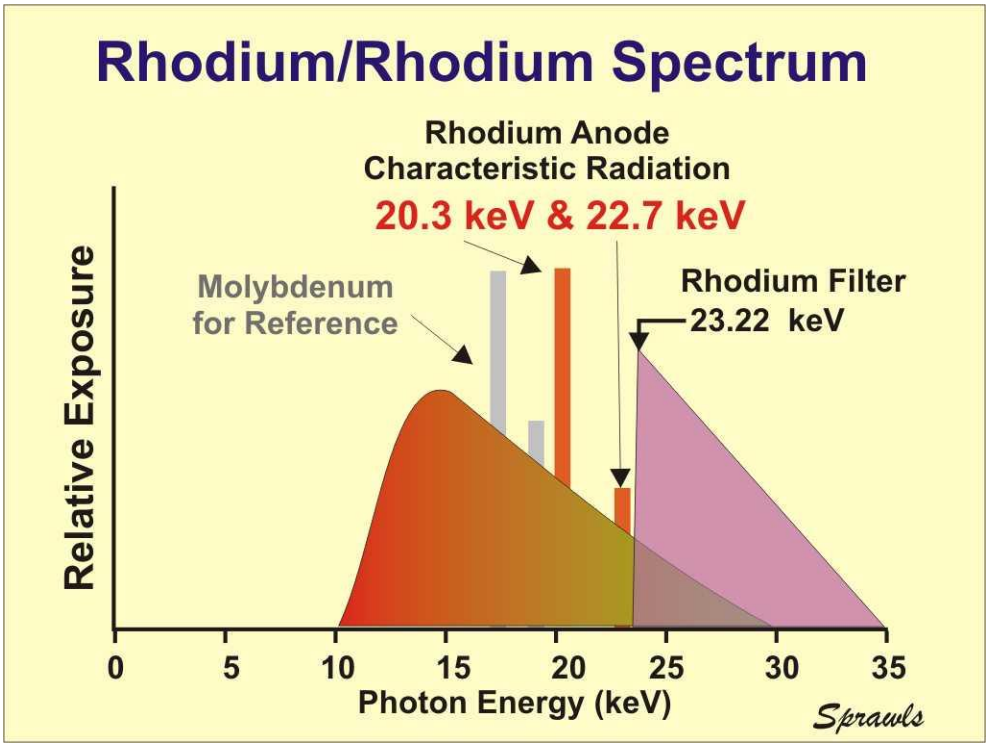


Figure 11. The spectrum produced with a rhodium anode and filter.

The advantage of rhodium over molybdenum was higher photon energy and a more penetrating x-ray beam. This had the effect of reducing doses to patients and potentially better visualization through some denser breast tissues. This was developed and applied in two phases, the filter and the anode.

A rhodium filter is a relatively simple small metal object that can be added as an alternative filter to the many systems with molybdenum anodes. It can then be used with the molybdenum anode tubes to increase penetration by passing the *bremsstrahlung* between the energies of 20 keV and 23.22 keV, the differences between the two K edges.

In 1992 General Electric introduced a dual track x-ray tube with a molybdenum anode track, molybdenum filter, a rhodium anode track, and a rhodium filter. The operator could select the function to produce either the spectrum shown in Figures 10 or 11, depending on the characteristics of the breast being imaged.

VI. DR. CHARLES MARIE GROS, SENOLOGY, AND THE SENOGRAPH

Dr. Charles Marie Gros was a physician and physicist serving as Professor of Medicine from 1950 – 1975, and Head of the Department of Radiotherapy and Radiology at University of Strasbourg, France. In 1963, he created a multidisciplinary medical specialty for the care of breast diseases and established the term *Sénology*. In his landmark publication by that title he defined senology (the study of the breast) as a neologism derived from the Romanic “seno” and the Greek “logos” as the branch of knowledge concerned with the mamma and the breast.

In 1975 he founded the Société Internationale de Sénologie (SIS) and published the first Journal on Breast Diseases: *Senologia*.

He developed the first equipment exclusively dedicated to breast imaging and collaborated with the Compagnie Générale de Radiologie (CGR), in developing and promotion of the mammography equipment called the Senograph shown in Figure 12.

Figure 12. The contributions by Dr. Gros included the formation of a society and journal and the development of the first dedicated breast imaging system, the Senograph.

VII. THE TRANSITION TO DEDICATED MAMMOGRAPHY SYSTEMS

In 1965 the Senograph developed through the efforts of Dr. Gros in collaboration with "Compagnie Générale de Radiologie (CGR)" became the first dedicated breast imaging system. One of its major features was using molybdenum for the anode and filter.

It is reported that by 1970 CGR had sold approximately 2000 Senographs throughout the world. Later General Electric purchased CGR and continued with Senography as the brand name for its dedicated breast imaging systems.

Other manufactures developed dedicated systems. These include the Mammomat by Siemens, MicroDose by Philips, and several brands by Hologic. Each of these could have included some special features but also the common features optimizing them for breast imaging including:

- Low and Adjustable KV
- Molybdenum Anode and Filters
- Grids with Low Attenuating Interspaces at the Low Photon Energies
- Dual Small Focal Spots
- Ability to Rotate for Different Anatomical Views
- Breast Compression and Positioning Capability
- Automatic Exposure Control Selectable by Operator

Before the transition to digital, the receptors were not provided by the equipment manufactures but by the major film industry including: Kodak, DuPont, Fuji, and Agfa. With the exception of xerography as described later, there were two major receptor components, *film* and *intensifying screens*. Each of these progressed through extensive developments contributing to improved image quality and controlling radiation dose to patients.

VIII. THE EVOLUTION OF FILM AS A MAMMOGRAPHY RECEPTOR

The special image quality requirements for mammography, especially high contrast sensitivity and visibility of detail, could not be provided with conventional radiography receptors. Receptors specific for mammography have been developed and have evolved throughout history. There are several desirable receptor characteristics that have motivated and guided the continuing innovations and development over the years.

Contrast Sensitivity and Dynamic Range

The transfer of relatively low physical contrast (differences in tissue densities) to visible contrast in images is one of the major challenges in mammography. This is determined by the contrast sensitivity of the imaging system. The first step in meeting this is through the optimized x-ray spectrum that has been described. When the invisible x-ray image from the

breast is delivered to the image receptor there continue to be several factors that determine the contrast that will be visible in the final image.

A related factor is the range of x-ray exposure to the receptor over which contrast will be produced. This characteristic is the *latitude* for film and *dynamic range* for digital receptors. There is generally a conflict between high visual contrast and latitude. This has resulted in the design of film specific for mammography and the necessity of special viewing conditions to be used by physicians.

Film with Direct X-ray Exposure

When mammography was being developed general radiography for all other parts of the body was being conducted with film exposed in cassettes with intensifying screens. This was not satisfactory for mammography for two major reasons. The blurring from the intensifying screens did not provide the adequate visualization of the small calcifications, and the film latitude (dynamic range) could not produce the necessary contrast over the wide range of receptor exposure caused by the variation in breast thickness. These two limitations were overcome by using film exposed without intensifying screens. The first mammographic receptors developed by Egan were film exposed directly by the x-radiation without intensifying screens as illustrated in Figure 5. The preparation of the film for imaging is shown in Figure 13.

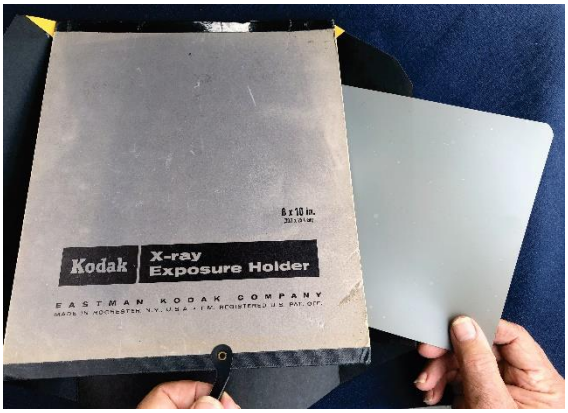


Figure 13. Industrial type film being inserted into a cardboard holder to be used as a receptor for mammography.

A specific type of film selected by Egan was Kodak Type M designed for industrial radiography. It had fine-grain, high density, and a thick emulsion. The thick emulsion required longer processing times than conventional radiographic film of the time. This was provided with either modified film processors or manual processing as shown in figure 15. The manual processing was recommended for maximum image quality.

Future developments in film and intensifying screen technology led to the replacement of directly exposed industrial film with intensifying screen-film combinations designed specifically for mammography. This occurred in two major phases and the film characteristics for each are compared in Figure 14.

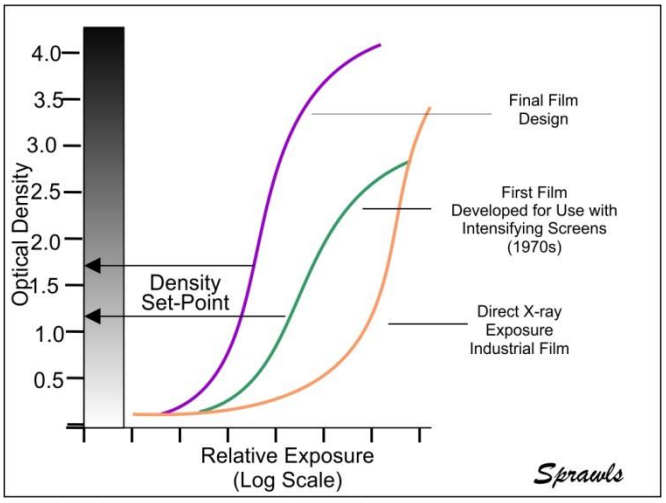


Figure 14. Characteristic (H & D) curves of the three major phases of film used for mammography.

This illustration compares the features of each type beginning with the directly exposed film. A major difference between films exposed with x-radiation and light is that with light exposure there is a limited maximum optical density that can be achieved resulting in the shoulder on the characteristic curve. With the direct x-ray exposed films to obtain the wide latitude required long chemical development times and manual or hand processing. With some of the image contrast recorded in the high-density or dark regions of the film special bright lights were required for viewing. These two characteristics are illustrated in Figure 15.

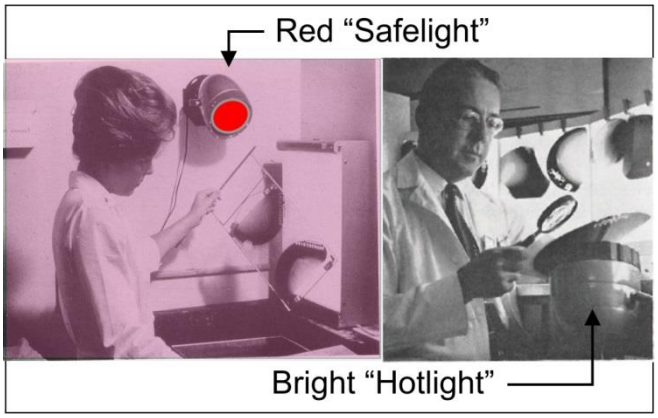


Figure 15. The manual processing of mammography film and Dr. Egan viewing the high-density areas with a bright "hot" light at Emory University.

Automatic film processors were used for general radiography but were not adequate for processing the industrial type film used in mammography. This film had a thicker emulsion and required longer times in the developer solution to convert all of the exposed grains to opaque optical density. The film was not sensitive to the red region of the light spectrum and the darkroom could be illuminated with a red "safelight" to provide visibility of the process. It required considerable experience and skill to work with the many variables associated with the process.

This was the beginning of the necessity to view mammography images with a bright light because of the extended density (opaqueness) into the darker range to provide the wide latitude/dynamic range, especially needed because of the variation in breast thickness as illustrated in Figure 8. The necessity for special "bright light" viewing returned as an issue years later with the design of the film for mammography that had extended latitude into the more dense (opaque) regions as illustrated in Figure 14 for the final film design.

Direct X-ray Exposed Film Summary

The directly-exposed industrial type film as a receptor for mammography was a major factor in the development and evolution of modern mammography. The ability to produce images with high visibility of detail (calcifications) and adequate contrast over a wide exposure range made effective clinical mammography possible, especially with the Egan Technique. One of the several challenges was that the manual processing of the film required considerable effort.

The characteristic that was the major concern and driving force for the development of other receptors was the high x-ray exposure required to form images when the film was exposed directly by the x-radiation.

This motivated the development of intensifying screen – film combinations for mammography. This generally occurred in two phases relating to developments of intensifying screens for all radiographic procedures. Films with specific characteristics to be used with each type of intensifying screen were developed. It is the film within the receptor that determines the contrast characteristics of an image and the contrast sensitivity of the imaging procedure. Before considering the intensifying screens we follow the evolution of film characteristics, and impact on visibility using the illustration in Figure 14.

Film Used With Intensifying Screens

A general characteristic of film exposed with light, compared to direct x-ray exposure, is a lower limit to the maximum optical density, or opaqueness that can be achieved. This has an impact on image contrast and especially the range of exposure (latitude or dynamic range) over which adequate contrast and visibility can be developed. A major factor relating to variations in breast thickness was described previously.

The receptors for mammography with intensifying screens used one screen rather than the two screens used for most other radiographic applications. This was to reduce image blurring as will be discussed later. The film had an emulsion on one side of the film base and was used with a single intensifying screen. However, the film emulsion was thicker for the purpose of producing an optical density comparable to films with emulsions on each side. The thicker emulsions generally required longer times for the chemical development to reach completion and special automatic processors with extended processing times were sometimes used.

The contrast characteristics of these earlier mammography films were not extensively different from film for other radiographic procedures. The films were exposed to produce approximately same range of densities as other radiographs and viewed under similar conditions.

The third and final phase of film-based mammography receptors resulted from innovations in both the intensifying screens, to be described later, and the film described now. The contrast characteristics associated with these three phases are compared in Figure 14 and will be used here as a reference.

A continuing objective in film design was to provide necessary image contrast over a wide range of exposure to the receptor--that is, wider latitude or dynamic range. With radiographic film one factor that limits latitude is the maximum optical density that can be produced. The specific characteristic that contributed to the wider useful latitude is the film emulsion design that can produce higher optical densities, or so called " D_{max}".

To benefit from this film design required two changes in practice. First, the images needed to be exposed to a higher average optical density to fall within the wider latitude. The automatic exposure control was calibrated for a density set point of approximately 1.7 compared to approximately 1.2 for the earlier film types as illustrated in Figure 14. This was the film density that resulted from exposing a phantom test object of uniform thickness representing an average breast. The second factor was that the denser or darker images required different and special viewing conditions.

Mammography Film Viewing

The viewing conditions for mammograms recorded on the two film types are compared in Figure 16.

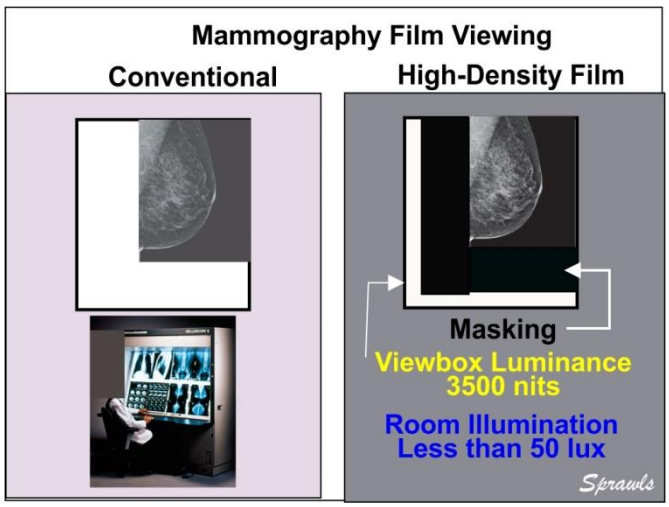


Figure 16. Comparison of the viewing conditions for the two general types of mammographic films.

For years the mammography film used with intensifying screens produced approximately the same optical density images as other radiographs and was viewed under the same conditions illustrated above. However, with the development of film producing images with greater optical density (opaqueness) the conventional viewing conditions were not adequate. The human visual process could not see all of the contrast and details in the darker images, and especially with surrounding glare and bright viewing rooms.

This was overcome with viewing conditions specific for mammography with three features. This included brighter illuminators (view boxes), masking around the small images while viewing, and darker rooms. The specifications, as required by some accrediting and regulatory organizations, are shown in Figure 16.

The Chemical Processing of Film for Mammography

The chemical processing and development of films for mammography had special requirements. For some types of film this was extended development times to achieve increased density and contrast. However, the trend was to design mammography films that could be processed along with other radiographic films in automatic processors that were the standard at the time. A major challenge was that film development was a chemical (not physical) process and subject to many variables including type and quality of the chemistry, replenishment as it was used, and solution temperature. Even in an automatic processor it was potentially an unstable and varying process. What varied was the level of development that determined how many of the exposed silver halide grains (the invisible latent image) were converted to visible density in the final image. This affected both the sensitivity and contrast characteristics of the receptor. The concern with under-developed film was both a loss of contrast and the requirement of higher exposure. Variation in development levels (consistency) could contribute to exposure errors and the necessity of repeating examinations.

Quality Control (QC) procedures specific for mammography film processors became a recommendation and requirement in many countries. These were often under the direction of medical physicists.

IX. XEROMAMMOGRAPHY

The xerography process, from the Greek, *xeros*, "dry" and *graphia*, "writing" was developed and used extensively in equipment for making copies of documents and images. Unlike other methods, including photographic film, it does not use "wet" chemicals but a completely dry electrostatic process to form images. The basic process as used for mammography is illustrated in Figure 19.

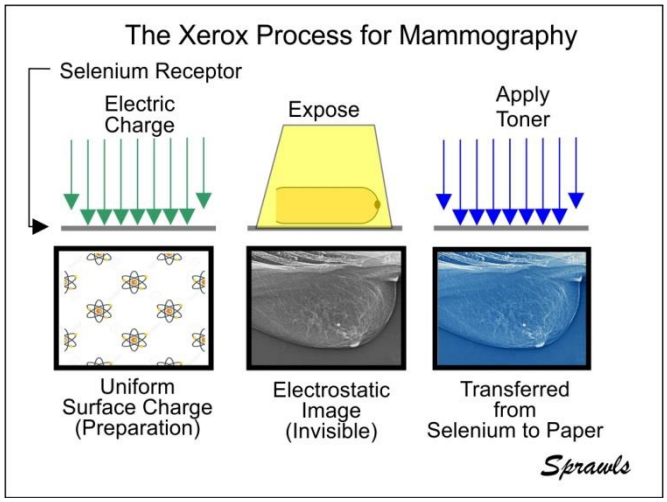


Figure 19. The three major steps in the formation of a Xerox mammogram.

The active component of the receptor is a layer or plate of selenium, an electrical semiconductor, enclosed in a lightproof cassette for imaging.

Preparation

In the first step within the processor the plate is cleaned from previous use and an electrical charge applied to the surface. It is then re-inserted into the lightproof cassette and ready for imaging.

Exposure and Image Formation

When exposed to x-radiation the selenium plate becomes conductive and discharged at each location in proportion to the exposure. This forms an invisible image in the form of a variable electrical charge on the surface of the plate.

Processing and Image Development

The cassette is inserted into the processor where the selenium plate with the electrical charge is sprayed with a fine-grain blue powder or "toner". The powder collects at each point on the surface in relationship to the charge and forms a visible image. It is then pressed onto a sheet of paper transferring the image. With some additional processing and sealing it is expelled from the processor as a permanent printed image.

Characteristics of Electrostatic Images

It was the unique way that electrostatic images, as different from chemical photographic images, are formed that provided several advantages for mammography. A major characteristic is that the attraction of the blue powder toner is most prominent at local transitions or gradients in the electrical charge and less dependent on the actual charge value throughout the larger image area. This produced images with two very valuable characteristics for mammography.

Edge Enhancement and Wide Latitude

A major challenge in breast imaging has been the large variation in breast thickness and density that could extend exposure beyond the latitude or dynamic range of film as described early. Another challenge is the need to image very small calcifications and anatomical detail. The unique characteristics of electrostatic imaging in the xeroradiography process provided solutions to both of these challenges as illustrated in Figure 20.

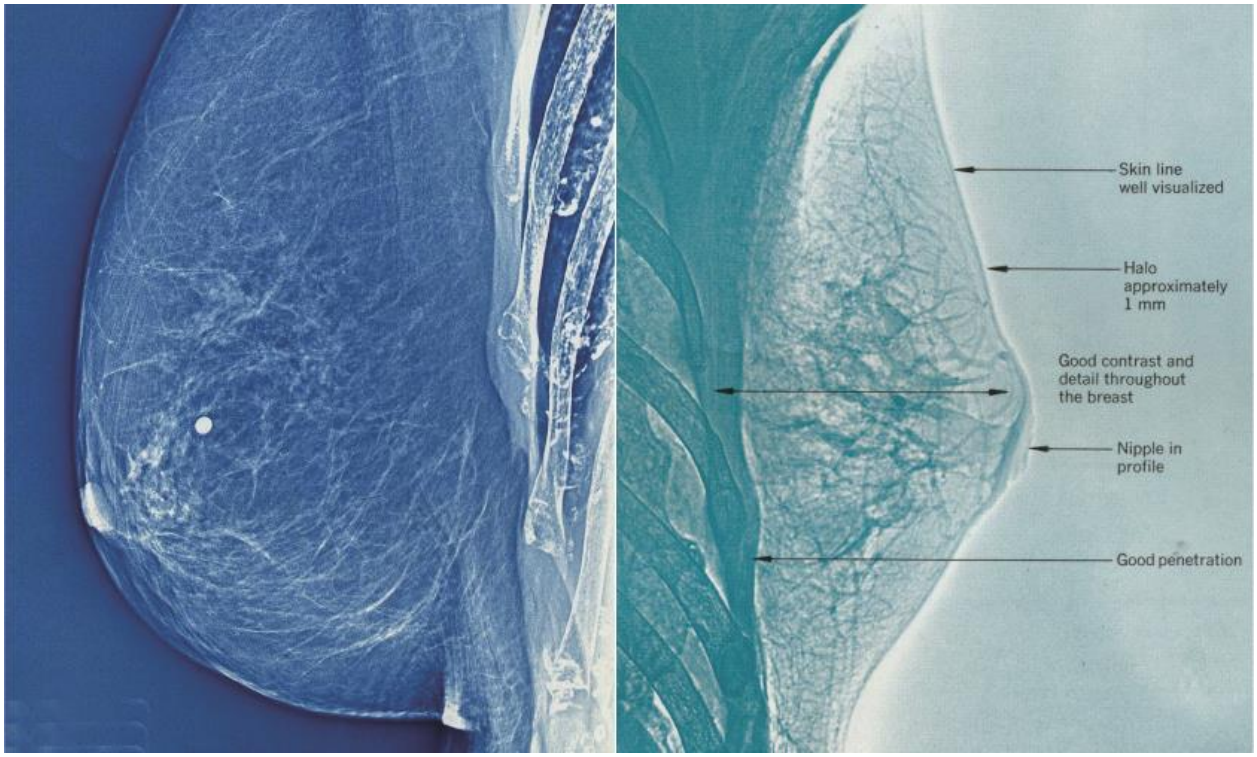


Figure 20. Images of a breast illustrating some of the advantages in technical brochure published by Xerox.

The edge-enhancement characteristic contributes to increased visualization of detail and small objects because they are more in the form of closely spaced edges or boundaries than being large areas.

Xerox for mammography was commercially introduced in 1971 and became a desirable alternative to directly-exposed film, both because of image characteristics and less radiation exposure compared to directly-exposed film. The images were especially appealing to physicians because of enhanced visibility of anatomical structures and the radiation dose was less than with directly exposed film. 1n 1985 black liquid toner was introduced but this did not contribute to the significant continuation of Xerox mammography.

As receptors with intensifying screens (LoDose, MinR, etc.) were developed and becoming widely used in the late 1970s their image quality characteristics and significantly lower radiation dose requirements contributed to the decline of xeromammography with commercial production ending in 1989.

X. THE INTRODUCTION AND EVOLUTION OF INTENSIFYING SCREENS

Mammograms produced with film as the receptor and exposed directly by x-radiation had good quality. The good contrast characteristics and low blurring was a major factor establishing mammography as a highly valuable procedure for diagnosing breast cancer. However, a major concern was the high exposure required to form images. This was a motivation to develop receptors with intensifying screens. Film-screen combinations, either in cassettes or rapid film changers were the receptors used in virtually all radiographic procedures. Even the so-called “detail” screens produced higher blur than was needed to image the small calcifications. The standard design of film-screen receptors used a film with the emulsion on both sides of the film base “sandwiched” between two intensifying screens. With these receptors there were three sources of blurring that limited visibility of detail (including calcifications) and making them not appropriate for mammography. These were 1. The thickness of the screens necessary to provide x-ray attenuation; 2. light crossover through the film base between the two emulsions; and 3.some possible space between the film and screen surfaces because of problems with film-screen contact. It was necessary to address each of these factors in the design and application of intensifying screens for mammography. The characteristics of intensifying screen receptors for mammography are compared to those for general radiography in Figure 21.

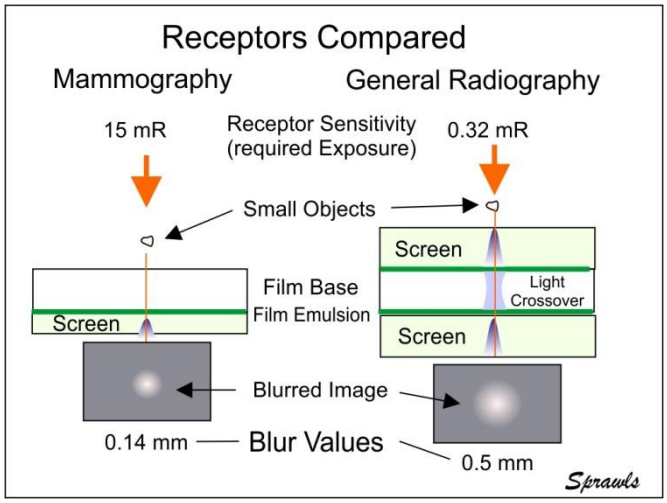


Figure 21. Comparison of intensifying screen receptors for mammography to general radiography.

The delay in using intensifying screens for mammography was the inherent blurring that limited visibility of calcifications and related anatomical detail. The two specific design features to address this was using a thinner screen and only one screen combined with a film that had emulsion on one side…different from conventional radiography that uses two intensifying screens with the emulsion on both sides of the film. Placing a film into a mammography cassette is shown in Figure 22.

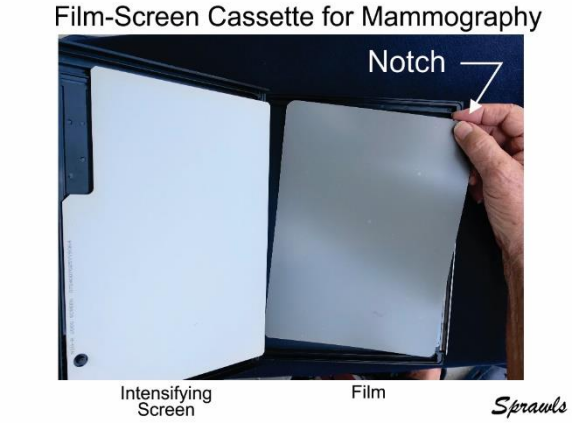


Figure 22, Inserting film into a single screen mammography cassette. This was in a darkroom so there was a notch in the film to identify the upper right corner when viewing the emulsion side.

While the development and transition to intensifying screens for mammography was motivated by the need to reduce radiation dose, the resulting exposure requirements would continue to be much higher than general radiography. In mammography there are two major factors that determine radiation dose to the breast, and both are in conflict with requirements for high image quality. One is the x-ray beam spectrum that should be optimized to balance dose with requirements for high contrast sensitivity, as described previously. The other is the exposure required by the receptor to form an appropriate image where increased exposure is required to reduce both blurring and visual noise.

Receptor Sensitivity (Speed) and Exposure Requirements

Transition from directly exposed film to Xerox mammography and then to intensifying screens provided a reduction in the exposure required to form images. However, the reduction in exposure and dose continued to be limited by image quality requirements, especially the effects of blurring and visual noise. The thin intensifying screen used with a single emulsion film resulted in an equivalent blur value of approximately 0.15 mm compared to values of around 0.5 mm for general radiography receptors. While this low blur value enhanced visibility of detail, specifically small calcifications, it also contributed to increased noise. That is because blurring has the effect of integrating photons within an area and reducing noise. This results in receptor sensitivity (required exposure) values of approximately 15 mR for mammography compared to 0.32 mR for a typical 200 speed general radiography receptor.

The receptor sensitivity values shown in Figure 21 are the input exposures required by receptors to produce a specific reference film density, generally one unit above the base plus fog density. It can be considered as an approximation of the average receptor exposure to form an image.

Because reduction in dose was the major motivation for developing intensifying screens for mammography this was emphasized in the early brand names including LoDose by DuPont and Min-R by Kodak.

In 1972 DuPont introduced the LoDose receptor that consisted of a thin calcium tungstate screen used with a single emulsion film enclosed in a flexible vacuum bag so that the earth's atmosphere pressed the film and screen together for good contact. To prepare for each image the technologist would insert the film into the bag with the intensifying screen and then use a manual vacuum to produce the compression.

In 1976 the rigid cassette was introduced which contained one screen and used with single emulsion film. This included the DuPont LoDose-2 continuing to use calcium tungstate and the Kodak Min-R system using a gadolinium oxysulfide screen. It was at this time that intensifying screens for radiography were transitioning from calcium tungstate that had been used for years to a variety of the rare-earth phosphors including gadolinium oxysulfide. Receptors with generally similar characteristics were provided by several other manufactures.

The evolution from calcium tungstate to the several rare-earth intensifying screens was a major advancement for general radiography. A contributing factor was the difference in atomic numbers (Z) between calcium tungstate and the rare earths. With the lower atomic numbers and K-edge energies the rare earths provided higher x-ray absorption rates within the x-ray spectrum used for general radiography and screen thickness could be reduced. Along with some associated developments the exposure to produce images was reduced. The transition to rare earth screens for mammography provided some reduction in exposure and dose but mammography remains a relatively high exposure procedure as will be discussed more below.

XI. VISUALIZING CALCIFICATIONS

Calcifications are one of the significant signs of some breast cancers. It is not just the presence of calcifications--many are benign--but the size, shape, configuration, and distribution that must be evaluated to diagnose cancer. Calcifications within the breast are generally divided as either *macro* or *micro* with 0.5 mm being a dividing point. It is the micro calcifications with dimensions less than 0.5 mm that are generally associated with cancer.

Test Objects and Phantoms for Visualizing Calcifications

Test objects or phantoms to evaluate the visualization of the small calcifications played a major role in both the development of mammography methods and the continuing quality assurance procedures conducted by medical physicists. Two of the earliest developed by Egan is shown in Figure 23.

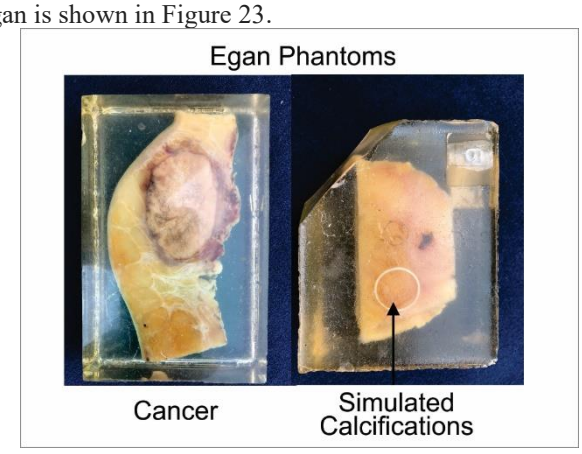


Figure 23. Phantoms developed by Egan to evaluate the contrast characteristics of a cancer and visibility of small calcifications in breast tissue.

The phantom was a section of breast tissue and cancer from surgery and embedded in a plastic block. Simulate calcifications with a range of sizes were added within marked circular areas. The image quality and visualization was evaluated by counting the number of calcifications visible with the circular area.

Over the years as mammography was being developed a variety of phantoms and test patterns were created and used. These included a design that was the standard for the accreditation of mammography facilities by the American College of Radiology (ACR) and various quality assurance procedures. A diagram and image is shown in Figure 24.

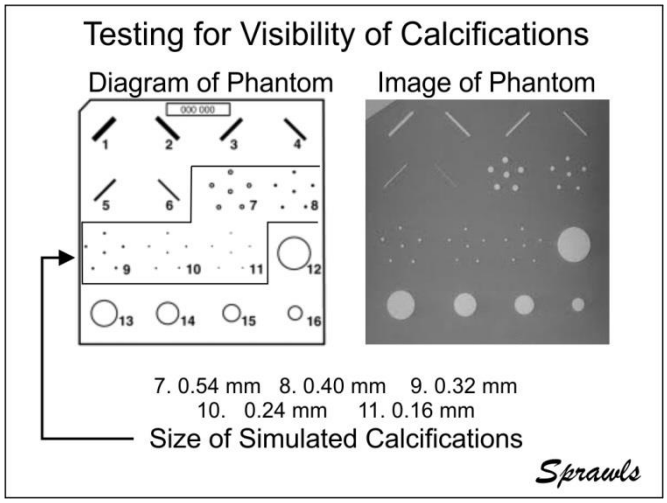


Figure 24. Diagram and image of the breast phantom that became a standard for evaluating mammography equipment performance.

The phantom was used to evaluate general contrast sensitivity with the larger circular objects that varied in thickness (physical contrast) and visibility of detail and calcifications with the "star shaped" clusters of simulated calcifications that are marked. Each cluster contains calcifications of specific sizes as shown. Image quality is evaluated by determining the highest-numbered cluster in which the calcifications can be seen. This gives a measure of the smallest calcifications that can be seen.

Blurring and Visibility of Detail and Calcifications

As with all imaging methods, it is the blurring within the imaging process that reduces and limits the visibility of small objects and detail. Effective mammography requires the ability to see the physical details, such as size and shape, of these micro calcifications. Therefore, mammography systems must be designed and operated to produce the least amount of blurring of any medical imaging process.

A general assumption or "rule of thumb" suggested by the author is that for an anatomical object to be visible in an image the dimension of the blur should not exceed the dimensions of the object. This is generally demonstrated in mammography. The effective blur values of typical mammography systems are in the range of 0.15 mm to 0.2 mm which is the approximate size of the smallest micro calcifications that can be visualized. Among all of the medical imaging methods, mammography is the one that requires the least blurring of all.

It is this requirement for very low blurring that has been one of the challenges and major objectives in the development, design, and operation of mammography systems over the course of its history.

Mammography System Composite Blur

The blur in a mammogram is the composite blur from the three major sources: receptor, focal spot, and motion. The source that produces the largest blur is generally the one that limits visibility and image quality. Because there are other factors associated with each source of blur, including focal spot heat capacity and the x-ray attenuation by receptors, there are limits to reducing these blur sources to very small values. In general, an optimized mammography procedure, with respect to blurring, is when the blur from the individual sources are about equal, unless one can be reduced more without compromises. Each of these has been addressed in the ongoing development of mammography.

The contribution of blur from each of the sources depends on the geometry or spatial relationships of components of the imaging system as illustrated in Figure 25.

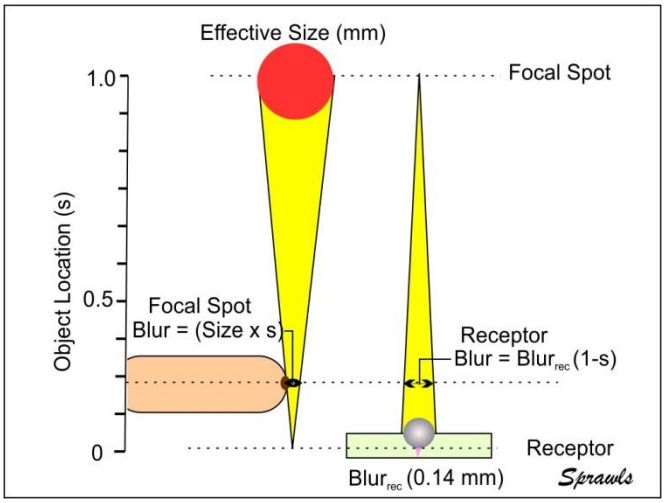


Figure 25. The effect of breast location ("s" scale) between the receptor and focal spot on the blur from those two sources.

The effect of blur on image quality and visibility of detail depends on the dimension of the blur in relationship to the dimension of an object. The geometric magnification within the imaging process is a factor in this relationship. Blur from the different sources is either magnified or minimized by the geometry as they relate to the location of the breast and objects to be imaged. Therefore, it is most appropriate to consider the dimensions of the blur at the location of the object. This relationship has been a major factor in the development of mammography over the years.

Effect of Breast and Object Location on Blurring and Visibility of Calcifications

The geometric configuration or spatial relationship of the focal spot, receptor, and breast for a mammographic procedure has been a major factor in image quality and has evolved along with other developments. The effect of breast location on image quality is best quantified using the "s" scale, rather than magnification factor as illustrated in Figure 25. This scale developed and published by the author (Sprawls) shows the location of the object being imaged, the breast, as a proportional distance between the receptor and the focal spot. The scale ranges from a value of "0" at the receptor to a value of "1.0" at the focal spot. Using this scale the value of the blur at the location of the object, where it directly relates to the size of the object, (calcifications, etc.) is a linear function of object location as shown in Figure 25. This applies to both focal spot and receptor blurring.

Receptor Blur

The necessity for reduced blurring was the major factor for selecting directly x-ray exposed film rather than intensifying screens in the early development of mammography. Intensifying screens were used in general radiography at that time but their inherent blur, and when used with double-emulsion film and some light crossover, were not adequate for mammography.

The development of intensifying screens specific for mammography in the 1970s was a major evolution because it provided for a significant reduction in radiation dose to the patient. The features of intensifying screens for mammography is compared to general radiography were compared in Figure 21.

XII. X-RAY TUBE FOCAL SPOTS FOR MAMMOGRAPHY

The x-ray tube focal spot has been a continuing challenge in mammography and has evolved with advances in technology in the quest for increased visibility, especially micro calcifications and anatomical detail. As we have just observed, the value and effect of focal spot blurring on visibility is determined by the combination of two factors, the effective size of the focal spot and the specific location of the breast between the receptor and the focal spot, along the "s" scale. Both of these have evolved over the years. This has included the development of tubes specific for mammography and dedicated equipment with generally fixed geometry.

Effective Focal Spot Size

The common practice in x-ray tube technology is to use two different quantities to express the size of focal spots. One is the *actual physical* or so called *"nominal" size* that can be measured by making images with a pin-hole camera or similar device. This is the size generally provided by the manufacturer within some relatively large tolerances. The *effective* (blur) size is what determines image quality. This is measured with star or line-pair resolution test patterns as often done by

physicists in quality assurance procedures. For a specific focal spot the effective value is always larger than the nominal because of several factors. The radiation distribution within the focal-spot area is not uniform--often two peaks because of the focusing characteristics of typical x-ray tube cathodes. This produces blurring as if it was a larger focal spot with either a uniform or Gaussian distribution of the radiation. A second factor is that the "nominal" size indicated on the label makes use of a tolerance factor so that it is actually smaller than the real physical size.

It is the effective size, as measured with test patterns that can be directly compared to the effective blur values of receptors to determine overall composite blur of an imaging system as it affects image quality. For each focal spot and receptor combination there is a breast location where the combined or composite blur has its minimum value. That is a significant factor in the design of mammography systems and has evolved over the years. With respect to focal spot size and system geometry there have been three major phases in the development of mammography as illustrated in Figure 26.

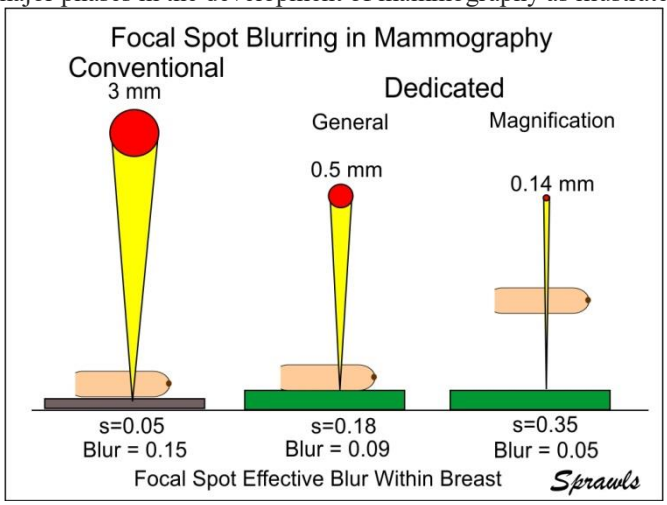


Figure 26. Three focal spot sizes that have been used in mammography.

Conventional X-Ray Tubes

In the early stages of mammography development conventional tubes with relatively large focal spots were used. They were available and the heat capacity of the larger focal spots was needed to produce high exposures in the shortest time possible with the directly exposed film. However, the geometry compensated for the large spots to minimize blurring. The focal spot-to-receptor distance was relatively long and the breast was very close to the film receptor.(s= 0.05). This was the combination that provided good visualization of calcifications and helped established mammography as a major method for detecting and managing breast cancer.

X-Ray Tubes for Mammography

With the clinical value of mammography having been demonstrated by Egan and others using conventional x-ray systems, often with some modifications, the motivation to develop x-ray tubes designed specifically for mammography was established. The various features of these tubes included special anode materials as described earlier. Also most systems placed the tube cathode towards the chest using the Heel effect to reduce intensity at the thinner side of the breast.

Here the attention is on x-ray tube focal spots. The tubes developed and used in more recent dedicated systems typically had dual focal spots with nominal sizes of 0.3 mm and 0.1 mm. The corresponding effective sizes (relating to blurring) are approximately 0.5 mm and 0.14 mm illustrated in Figure 26. Some designs, especially for the smaller focal spots, used a method that focused the electron beam on the anode in a more Gaussian pattern that contributed to smaller effective (blurring) sizes in relationship to physical (heat capacity) size. The smaller focal spots for mammography, compared to conventional radiography tubes, were possible because of a combination of factors. With the use of intensifying screens the exposure (anode heating) was reduced and with a smaller field of view a more favorable anode angle could be used.

Magnification Mammography

The ability to use geometric magnification to improve image quality and the visualizations of micro calcifications was a significant development in mammography. With the availability of small focal spots geometric magnification can be used to reduce the effective receptor blurring as illustrated in Figure 27.

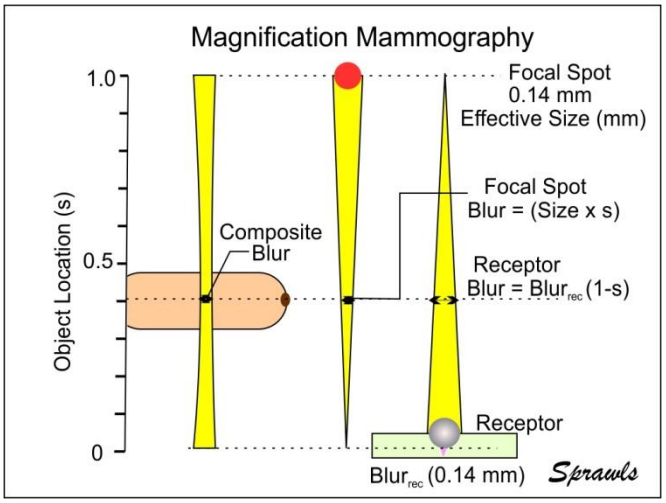


Figure 27. The use of geometric magnification to reduce the effective receptor blurring and increase visibility of small calcifications.

A major factor contributing to the value of this procedure is that the value of the composite blur from the two sources is less than the sum of the two values. It is a convolution or "overlap" of the two blurs. This advantage and improved visibility is achieved when focal spots with sizes smaller or close to the blur values of the receptors are available.

In 1977 Radiological Sciences, Inc. developed an x-ray tube with a very small focal spot size. One of the first units was installed in our laboratory at Emory combined with breast supports for performing magnification. A combination of physics and clinical investigations demonstrated the value of magnification to enhance visibility of detail and calcifications. Other researchers demonstrated the value of the magnification technique in extensive clinical studies. Dual focal spot tubes with a smaller spot for magnification as described previously became the standard for dedicated mammography systems.

XIII. RADIATION EXPOSURE AND DOSE IN MAMMOGRAPHY

The major goal in the continuing development and innovations in mammography was image quality and visibility within the breast. Its clinical value depended on that. A prevailing challenge was controlling the radiation dose to the breast. This was especially significant because of two factors: one is biological and one is physics. The biological is the relatively high sensitivity of breast tissue to undesirable biological effects from x-radiation compared to other anatomical regions. The physics issue is the dependence of several image quality characteristics on the quantity of radiation used to produce an image. With mammography requiring higher image quality than other radiographic methods it is inherently a high exposure procedure.

Breast Entrance Skin Exposure (ESE)

One of the activities often performed by physicists was determining the radiation exposure or dose to a patient. In the early period of mammography this was generally limited to determining the exposure delivered to the surface of the breast. This was done by calibrating the exposure output of the machine and then calculating from the technique factors (KV and MAS) actually used in a procedure. An alternative was placing TLDs on the breast.

The surface exposure was used to compare different methods and procedures but did not provide a dose value that was appropriate for evaluating biological risk. Some approximate surface exposure values for the different phases of receptor development are:

- Direct exposed industrial type film – 6,000 mR.
- Xeroxmmmography – 3,000 mR.
- Film with intensifying screens – 1,000 mR

These values illustrate two significant factors that mammography is a relatively high-exposure procedure and the required exposure was reduced with developments in receptor technology.

Mean Glandular Dose (MGD)

The concept of mean glandular dose (MGD) was developed as a quantity that would be more related to the biological effect of the radiation. It is defined as the average radiation absorbed dose to the breast glandular tissue and became the standard for monitoring dose in clinical procedures, evaluating equipment performance in quality control activities, and for some regulatory limits.

It is determined by multiplying the surface exposure value by a conversion factor that has been developed by physicists for a range of breast sizes and x-ray spectra characteristics generally specified with HVL values. For an average breast and typical imaging procedure a 1R SES will result in a MGD of 108 mR.

Two of the major factors that determine exposure and dose to a breast are the penetration characteristics of the radiation through the breast and the sensitivity, or required exposure, to the receptor. Both of these have evolved in the development of mammography.

Receptor Sensitivity and Required Exposure

Significant reduction in exposure and dose resulted in the transition of directly x-ray exposed film to the use of intensifying screens. Receptors for mammography continue to require a relatively high exposure to control quantum noise because of the very low blurring characteristic.

X-ray Penetration through the Breast

The effect of the x-ray spectrum on the opposing requirements for high contrast sensitivity (in soft tissue) and reducing radiation dose has been a major factor in the continuing developments in mammography. The goal is to adjust for an *optimized* spectrum for each breast size and density. A major contribution was the introduction of molybdenum and rhodium as anode and anode and filter materials. Compression and measurement of the thickness of the breast contributes to this effort. The training and experience of the mammographer / technologist is a critical factor in conducting an optimized procedure with respect to image quality and dose.

In addition to these two major factors that have evolved over time the introduction of grids and the magnification technique resulted in some increase in exposure and dose but are considered to be appropriate because of the increased image quality.

XIV. CHRONOLOGY OF DEVELOPMENTS IN MAMMOGRAPHY

The physics and technological developments to increase the clinical effectiveness, manage risks, and improve the overall efficiency of mammography have continued for well over a half century, and with more to come. It is appropriate to summarize by relating some of the major developments to the times when they occurred. This gives a valuable perspective to the scope of physics contributions to this medical specialty and the preservation of life and health for society around the world in specific decades.

The 1960s…The foundation

This was the period in which mammography began to be developed as a major medical procedure especially with the pioneering work and contributions of Drs. Robert Egan in the USA and Charles Marie Gros in France. Both were physicians but were major contributors to the application of physics and development of the technology for imaging the breast. By using the technology and methods developed under their leadership and in their collaborations with physicists and engineers they demonstrated, promoted, and expanded the clinical application through extensive educational and organizational activities. The technology at that time consisted of conventional x-ray equipment, tubes with tungsten anodes, and receptors consisting of industrial type film exposed directly by the x-radiation. The imaging procedure did not generally include compression and stabilization of the breast. In 1969 this decade was concluded with a major breakthrough, the development of the first dedicated mammography equipment that included a molybdenum anode and filter, the CGR Senographe.

The 1970s…Development of Modern Mammography Technology and Methods

The major physics and technology developments establishing mammography as a valuable and practical method for diagnosing and managing breast cancer occurred during the 1970s.

The first dedicated equipment, the Senograph, introduced in 1969, spread around the world along with dedicated systems developed by other manufacturers.

In 1973 the Siemens Mammomat and the Philips MammoDiagnost, Toshiba and Picker Mammorex were introduced. In 1974 General Electric introduced the dedicated MMX system.

In 1977 Radiologic Sciences Inc. provided a tube with a very small focal spot that stimulated the development of the magnification technique. This technology was acquired by Pfizer in 1979 and then by Elscint in 1981.

In 1978 Philips added an anti-scatter grid that was developed for mammography.

It was the decade for the development of image receptors for mammography to replace the industrial type film that required relatively high radiation exposures.

In 1971 Xeroradiography was introduced and was used for several years, generally the interval between directly exposed film and the development of intensifying screens specific for mammography.

In 1972 DuPont developed the Lo-dose calcium tungstate screen-film system contained in an evacuated bag to provide good film-screen contact during exposure.

In 1976 DuPont introduced the Lo-dose/2 screen-film system and Kodak the Min-R system using a rare earth screen. These were in rigid cassettes, much easier to use than the vacuum bags. Also, Agfa Gavert entered the market with a film screen cassette for mammography.

The 1980s…Refinements to Technology and Attention to the Total Mammography Operation

This decade began with mammography being performed with dedicated equipment and state of the art film-screen receptors. It was not to be a time for major innovations. Equipment features including automatic exposure control (AEC) were being refined. There were some advances in general radiography film design, the introduction of tabular "T" grain that was also used in mammography.

In 1987 the American College of Radiology (ACR) began its Mammography Accreditation Program, the ACR MAP. This was for Facilities that performed mammography. It was not a government legal requirement but some medical insurance providers would only pay for services in an accredited facility. There were a number of conditions required for accreditation including use of approved equipment and education of staff. A significant requirement was periodic quality control evaluations performed by qualified medical physicists.

This was the beginning of quality control procedures with specified image quality requirements, testing methods, and reporting that were to become a major role for medical physicists in mammography.

The 1990s…Image Quality Control and Personnel Qualifications

This was the decade in which emphasis transitioned from developments in new technology to the factors associated with the total mammography process including human performance. A major objective was to ensure that the high image quality available with the equipment and imaging procedures of that time was being achieved and contributing to accurate diagnosis and management of breast cancer. Medical physicists were to be a highly significant part of this development

In 1992 the American College of Radiology (ACR) published the *Mammography Quality Control Manual for Radiologists, Radiologic Technologists, and Medical Physicists.*
This provided detailed instructions and procedures for all …

In 1994 the USA Food and Drug Administration (FDA) implemented the Mammography Quality Standards Act (MQSA). This was a major action in which virtually all aspects of quality in mammography became regulated by federal legislation and law.

XV. AND THEN THERE WAS DIGITAL

The decades of the 2000s were to be the era of a major transition in mammography, from *film* to *digital* receptor, viewing, and image archiving technology and methods.

In 2000 the USA FDA approved the first digital system. The GE Senographe 2000D, for clinical use. Others were to follow.

The history of digital mammography is "another story for another day" and is not included here. The interests here are some of the factors associated with digital that brought to an end the use of film as the receptor element for mammography.

Photographic type film, a radiation sensitive emulsion coated on a transparent base, was, along with fluorescent intensifying screens, the foundation of radiography for over a century because of its many valuable characteristics. These included converting invisible radiation into visible images, an easy to view display, which could be stored and archived. However, along with these many values there were challenges and disadvantages that contributed to its replacement with digital technology. These included:

- An expensive silver based commodity that could be used only one time
- Required precise and accurate exposure to capture contrast from the breast.
- Required expensive, time and labor consuming, and somewhat unstable chemical processing.
- After being exposed and chemically processed images cannot be changed or adjusted, often requiring repeated exposures to correct for technique factor errors.
- Transporting and managing mammograms within a facility requires considerable time and effort.
- For an image on film viewing factors including brightness, contrast, and magnification cannot be adjusted.
- Archiving and retrieving images on film requires time and labor in addition to space with controlled environmental conditions.

Digital imaging technology provided solutions for all of these limitations, including:

- The wide exposure dynamic range of digital receptors that overcomes the prevailing latitude limits of film and related variations in breast size and composition.

- Images quickly transferred electronically from receptor to viewing display…without manual labor and chemical processing.
- Ability to control image viewing conditions to enhance visibility over a range of breast sizes and compositions.
- Electronic management, archiving, retrieval, and distribution of mammograms, almost "at the speed of light".

In addition to replacing and bringing to an end the use of film as a receptor digital technology made possible the development of the next major innovation in mammography, *tomosynthesis*.

And with that we conclude this history of the major phase in the development and evolution of mammography in which film served many valuable functions, from receptor element to display for viewing and archiving.

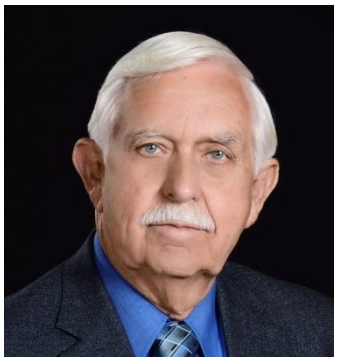
XVI. BIBLOGRAPHY

The historical evolution of the physics and technological developments in mammography has been researched and published by many who were often themselves active contributors to the ongoing activity. Their publications often contain details of specific developments beyond what is described in this article. They also provide extensive literature references to the science and clinical research contributing to the development of mammography.

- Gold, RH, Bassett, LW, Widoff, BE. Highlights from the History of Mammography (RSNA Exhibit) https://pubs.rsna.org/doi/pdf/10.1148/radiographics.10.6.2259767
- Bassett, LW, Gold, RH, The evolution of mammography. ***March 1988***, Vol. 150. No. 3 . American Journal of Roentgenology. 1988;***150: 493-498***. https://www.ajronline.org/doi/10.2214/ajr.150.3.493
- Vyborny, CJ, Schmidt, RA, Mammography as a radiographic examination: An overview Radiographics. Volume 9, Number 4 July, 1989 https://doi.org/10.1148/radiographics.9.4.2667052
- Haus, AG. Historical Technical Developments in Mammography *Technology in Cancer Research & Treatment (TCRT) Volume: 1 issue: 2, page(s): 119-126 Issue published: April 1, 2002* https://doi.org/10.1177/153303460200100204
- Haus, AG (1987) Recent Trends in Screen-Film Mammography: Technical Factors and Radiation Dose. In: Brünner S., Langfeldt B. (eds) Breast Cancer. Recent Results in Cancer Research, vol 105. Springer, Berlin, Heidelberg https://doi.org/10.1007/978-3-642-82964-2_6
- Haus, AG, Cullinan JE. Screen-Film Processing Systems for Medical Radiography: A Historical Review: Radiographics 9, 1989 https://doi.org/10.1148/radiographics.9.6.2685941
- Haus, AG, Technical Aspects and Image Quality in Mammography. AAPM 2002 AM https://www.aapm.org/meetings/02AM/pdf/8395-26604.pdf
- Paulus, DD**,** Imaging in Breast Cancer**,** Ca-A Cancer Journal for Clinicians. Vol. 37, No.3 May/June 1987 https://onlinelibrary.wiley.com/doi/pdf/10.3322/canjclin.37.3.133
- Gershon-Cohen, J. Breast Roentgenology: Historical Review. AJR 1961; 86:879-883.
- Joe, BN, Sickles, EA. The Evolution of Breast Imaging: Past to Present Radiology Vol. 273, No. 2S. 2014 https://pubs.rsna.org/doi/full/10.1148/radiol.14141233

XVII. ABOUT THE AUTHOR

Perry Sprawls. PhD, is a medical physicist and engineer specializing in diagnostic medical imaging. Mammography has been a major focus of his efforts including research and development, clinical support and procedure optimization, quality control and assurance, and extensive educational activities.



He joined the Emory University faculty in the 1960s and there was a collaborator with Robert Egan, MD, the "father of mammography" for many years in the continuing development of mammography technology, methods and applications. He gave special emphasis to the value of medical physicists working directly in clinical mammography facilities and in collaboration with radiologists and technologists to enhance image quality and related functions. In addition to his University and clinical activities he was a consultant and collaborator with several of the major developers and manufacturers of mammography image receptors. His mammography physics educational materials are available for all to use at: www.sprawls.org .

The collaborators who have been major contributors to his career in mammography are: Debra Monticciolo, MD, formerly at Emory, now at Texas A&M University Health Sciences; Earle Lee Kitts Jr., PhD, DuPont scientist; and Arthur G. Haus, FAAPM, Kodak scientist and medical imaging historian.

XVIII. ADDEDNIUM

Review of the Physics of Mammography

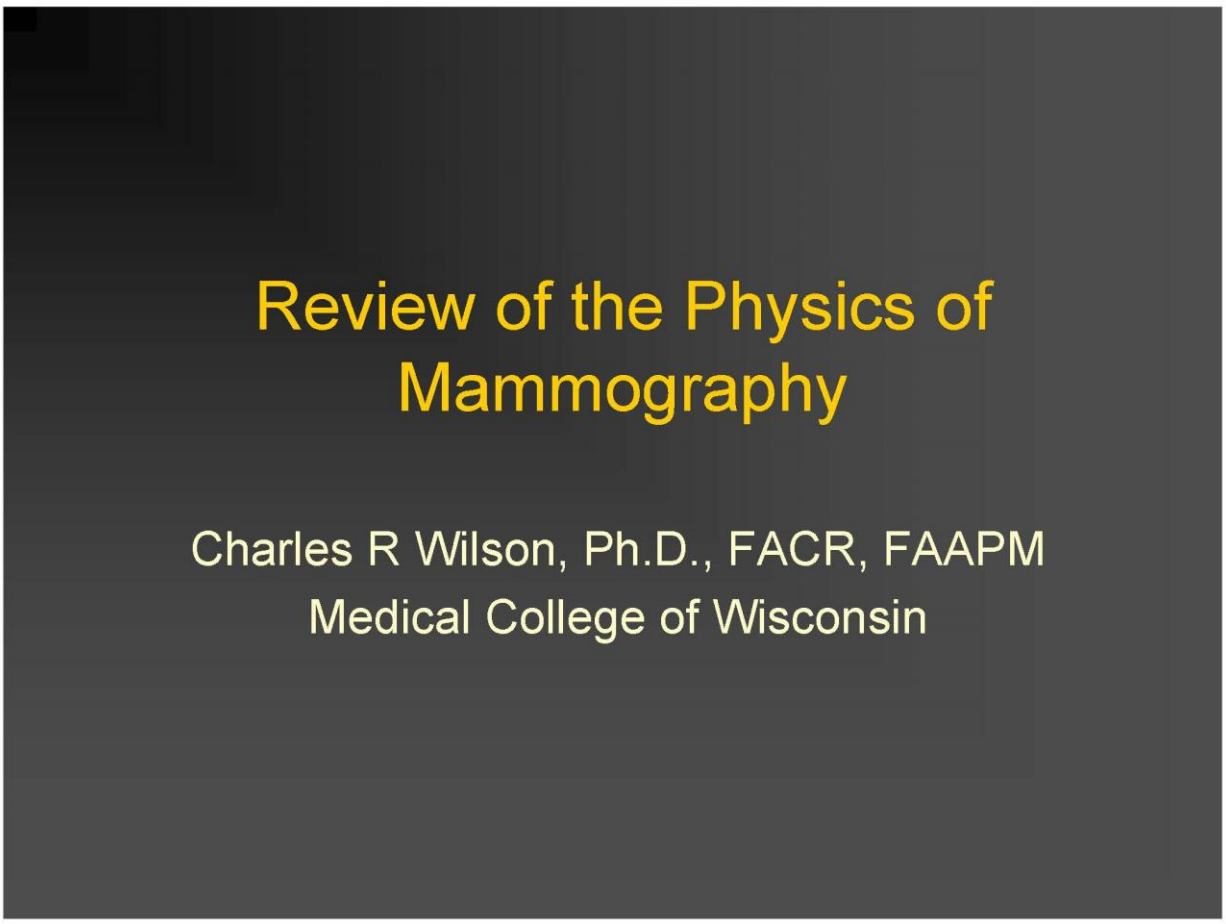
This course was presented at an American Association of Medical Physicists Annual Meeting. It provides an overview and extensive details covering the continuing developments of the technology and methods that document much of the history of mammography.

Charles R. Wilson, PhD, FAAPM, FACR
Medical physicist specializing in x-ray imaging
and related activities.

The contribution of Dr Charles R Wilson to this chapter of the Medical Physics History project is gratefully acknowledged.

Further down in the text follows the presentation of Dr Wilson as PDF in standard resolution.

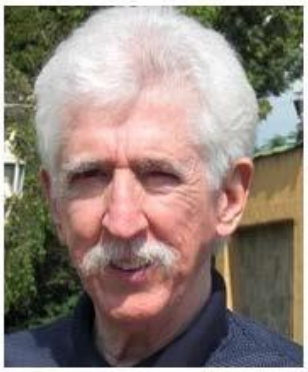
The High resolution PDF of this presentation can be downloaded here.

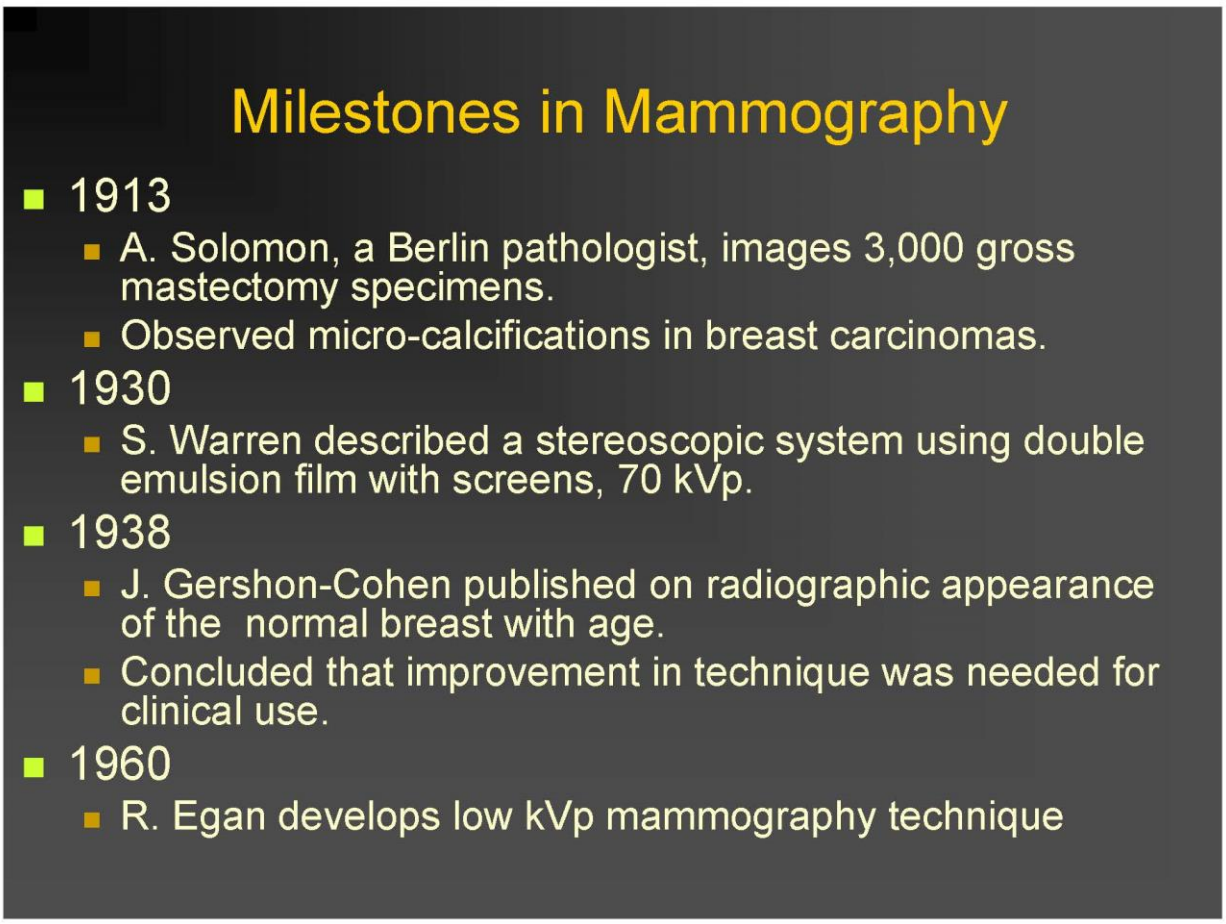

Review of the Physics of Mammography
Charles R Wilson, Ph.D., FACR, FAAPM
Medical College of Wisconsin

This course was presented at an American Association of Medical Physicists Annual Meeting

It provides an overview and extensive details covering the continuing developments of the technology and methods that document much of the history of mammography.

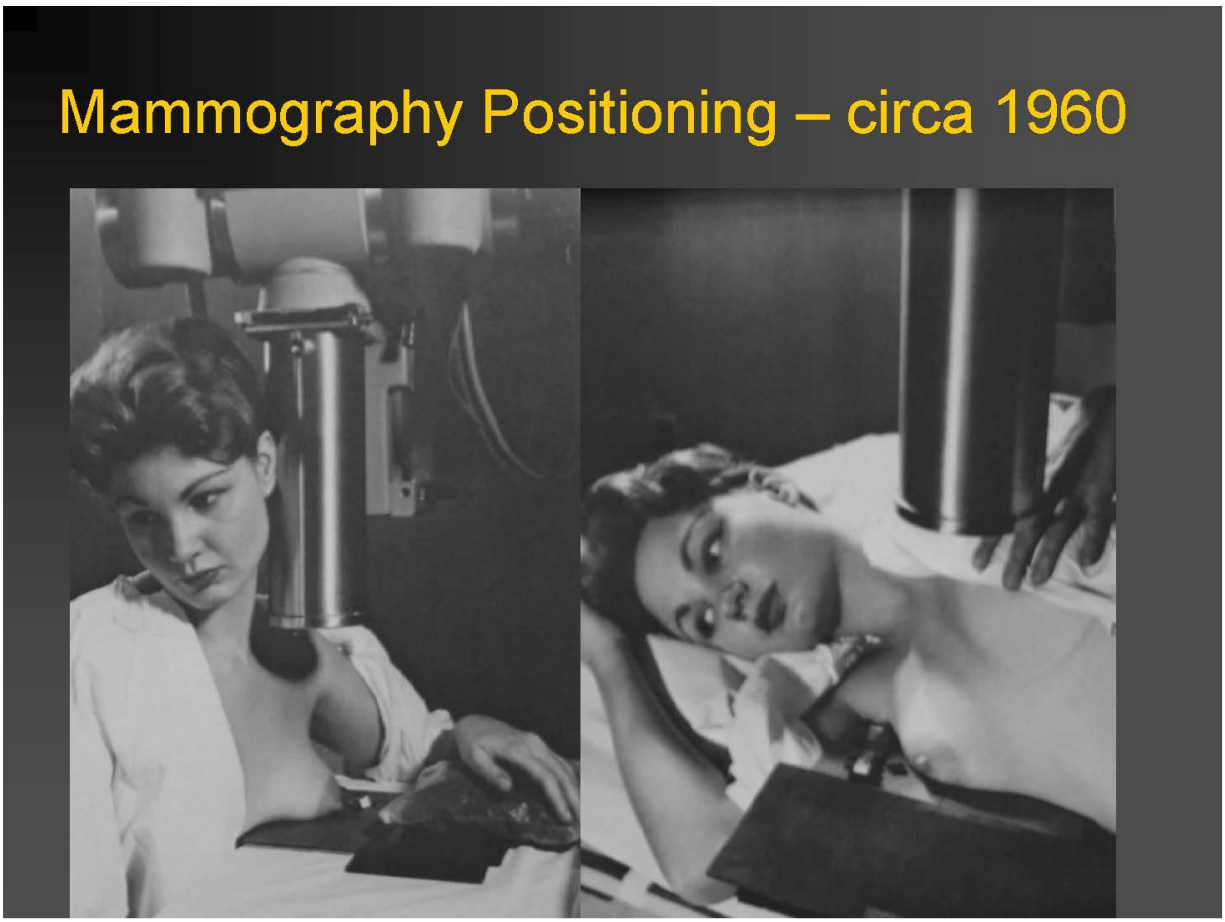
Charles R. Wilson, PhD, FAAPM, FACR
Medical physicist specializing in x-ray imaging and related activities at the
MEDICAL COLLEGE OF WISCONSIN

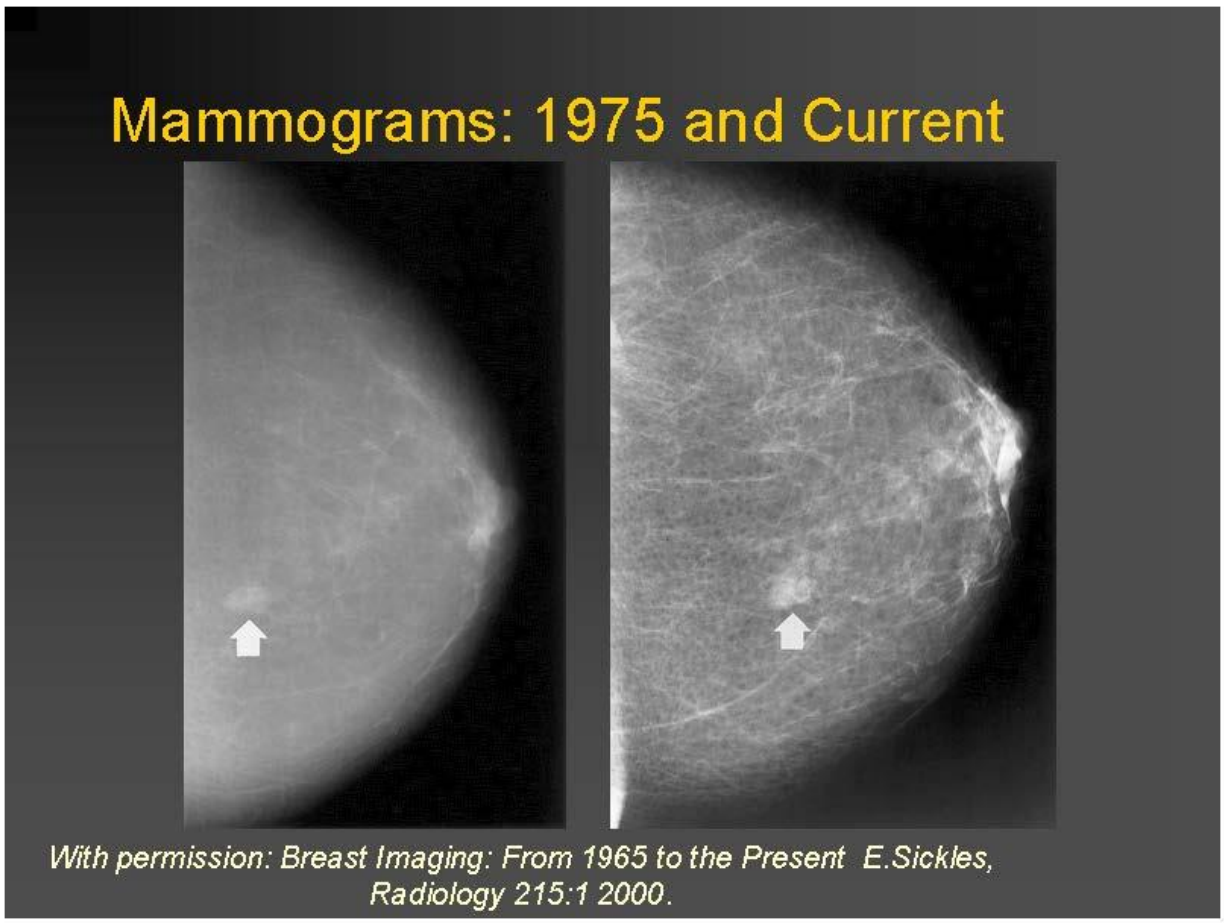


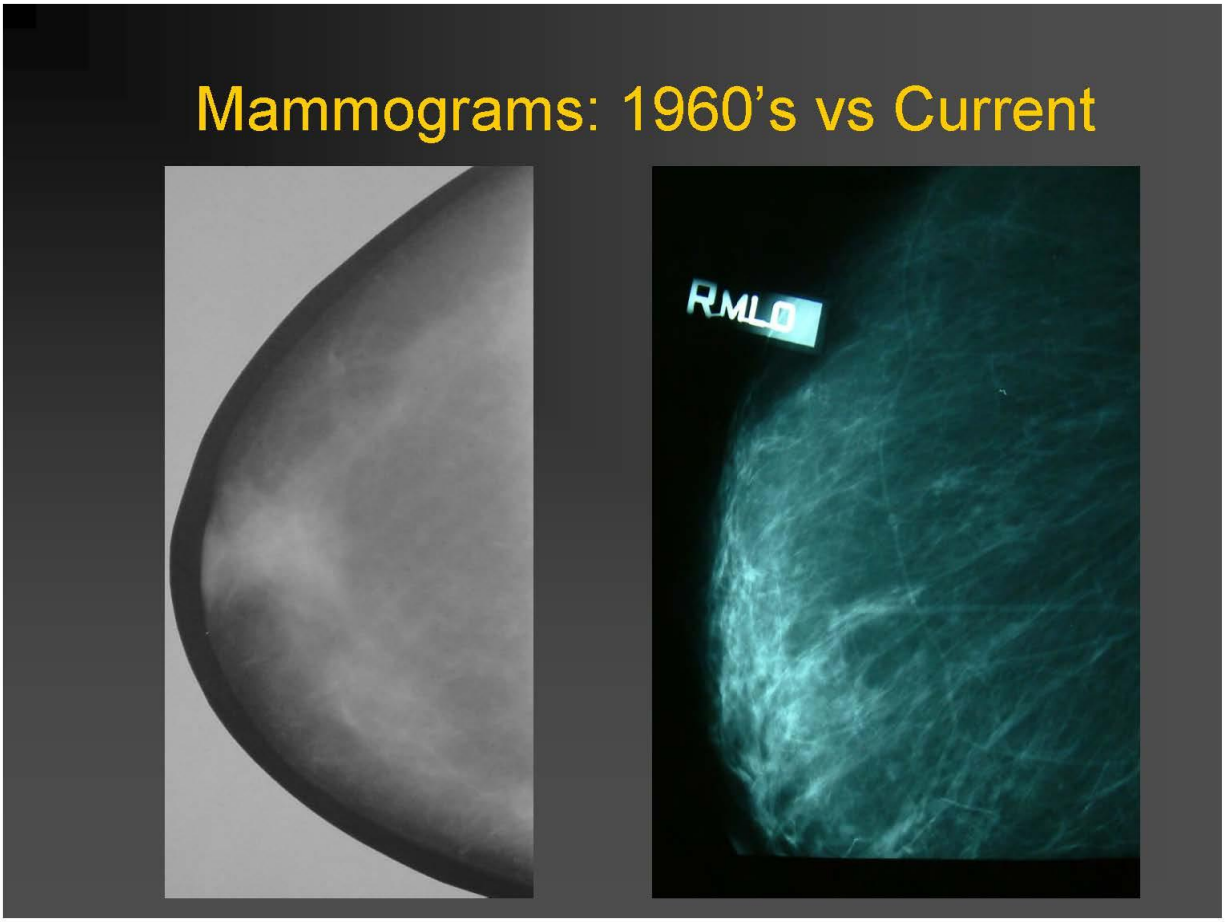

Milestones in Mammography
1913
A. Solomon, a Berlin pathologist, images 3,000 gross mastectomy specimens.
Observed micro-calcifications in breast carcinomas.
1930
S. Warren described a stereoscopic system using double emulsion film with screens, 70 kVp.
1938
J. Gershon-Cohen published on radiographic appearance of the normal breast with age.
Concluded that improvement in technique was needed for clinical use.
1960
R. Egan develops low kVp mammography technique

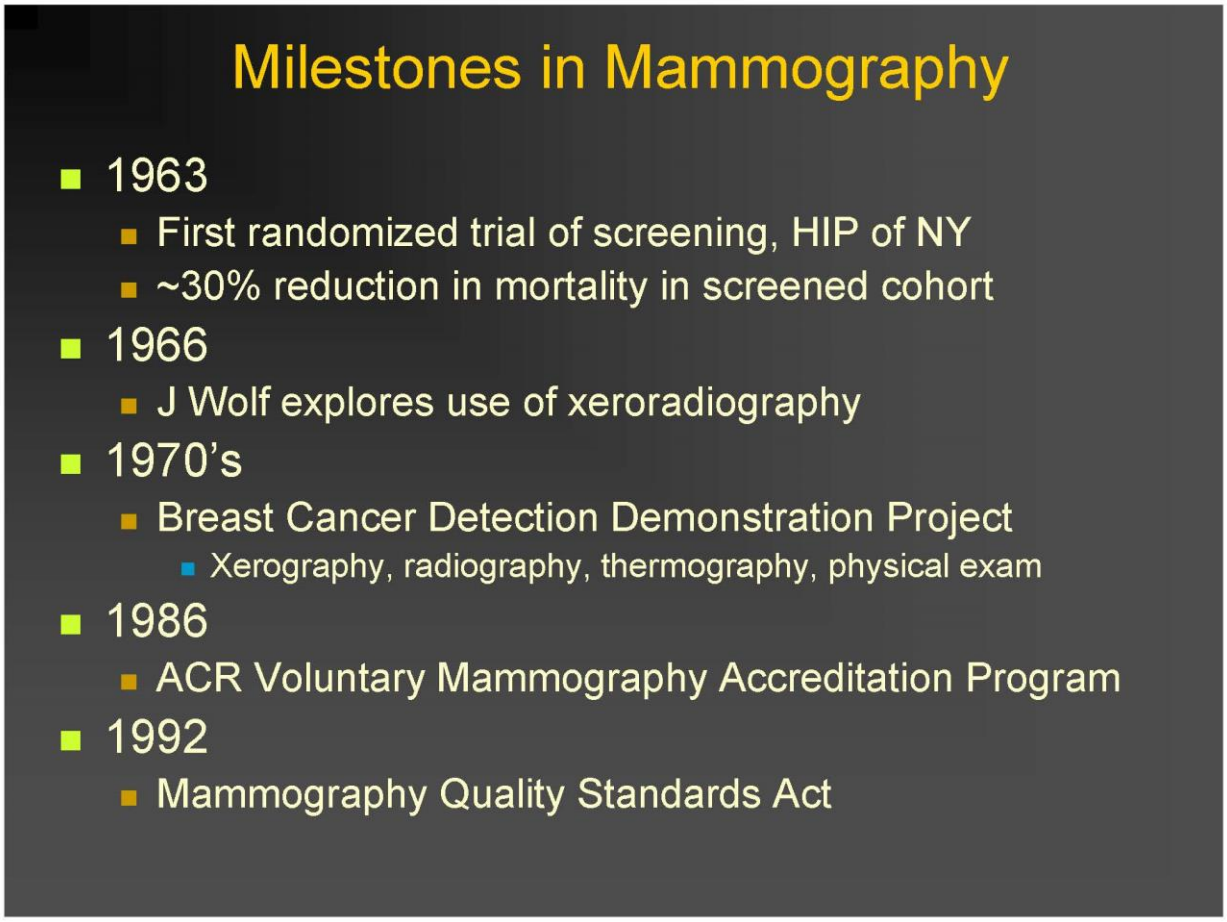
Robert Egan's Technique – 1960's

- Low kVp technique
  - verified kVp using a 15 mm Al wedge
- Beryllium window x-ray tube with minimum filtration
- Space charge limitations resulted in long exposure times, ~6 seconds
- Long SID: reduce focal spot blurring and provide adequate field coverage
- Metal extension cones: no field light
- Fine-grain industrial film
- No grid


Mammography Positioning – circa 1960

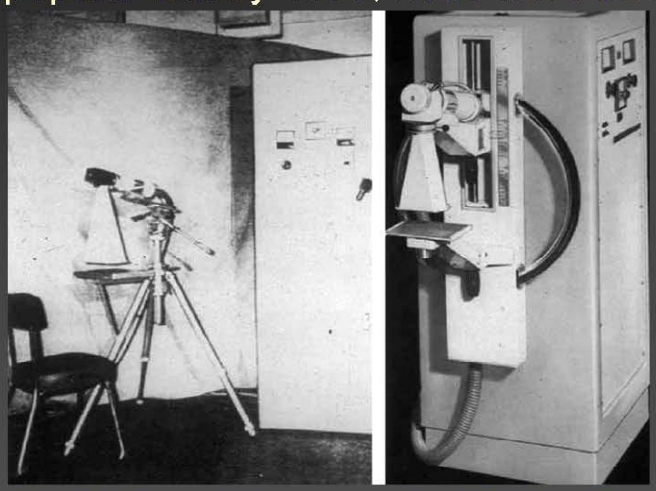

Mammograms: 1975 and Current
With permission: Breast Imaging: From 1965 to the Present E.Sickles,
Radiology 215:1 2000.


Mammograms: 1960's vs Current
RMLO


Milestones in Mammography
1963
First randomized trial of screening, HIP of NY
~30% reduction in mortality in screened cohort
1966
J Wolf explores use of xeroradiography
1970's
Breast Cancer Detection Demonstration Project
Xerography, radiography, thermography, physical exam
1986
ACR Voluntary Mammography Accreditation Program
1992
Mammography Quality Standards Act

First Dedicated Mammography Unit

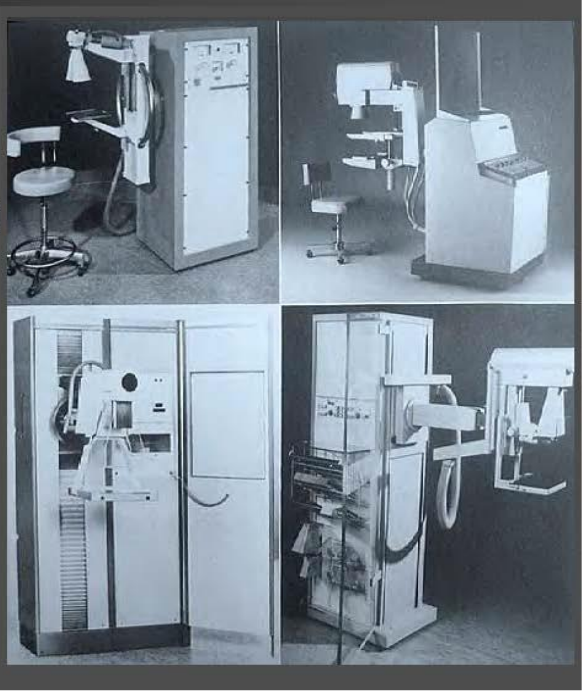
- 1965
  - Charles Gros, MD, Strasbourg, FR
  - CGR Senographe (Breast in French is Sein)
  - Very popular unit by 1970, 2000 were installed world-wide



With permission: Short History of Mammography: A Belgian Perspective, A. Van Steen, R. Van Triggelen, JBR-BTR, 2007.

Dedicated Mammography Units

- 1973
  - Picker (Mammorex),
  - Siemens (Mammomat)
  - Philips (Diagnost)
- 1974
  - GE (MMX)



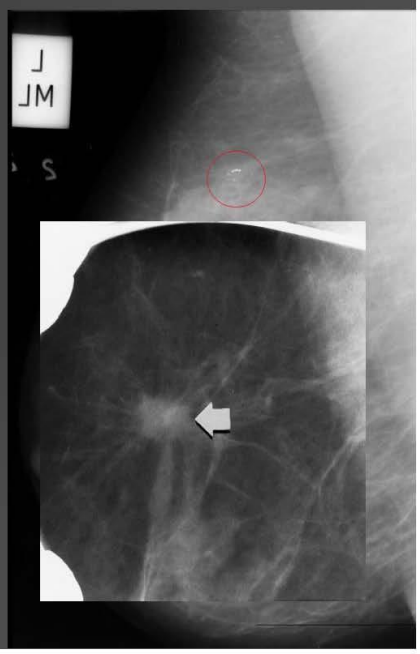
Current Dedicated Mammography Unit

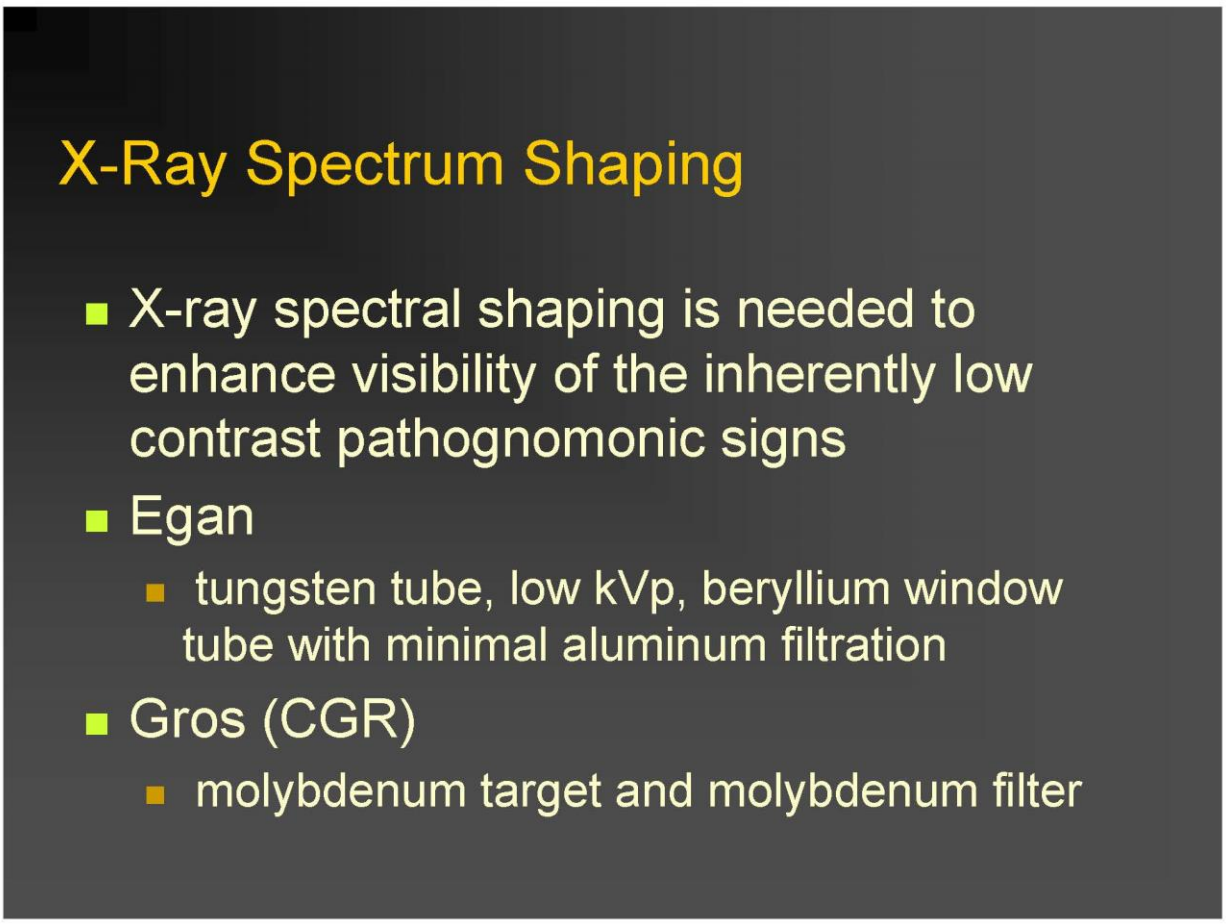
- Gantry mounted x-ray and detector assemblies
- X-ray tube target/filtration and focal spot appropriate for mammography
- Compression device
- AEC
- Film/screen and grid designed for mammography
- Dedicated film processor

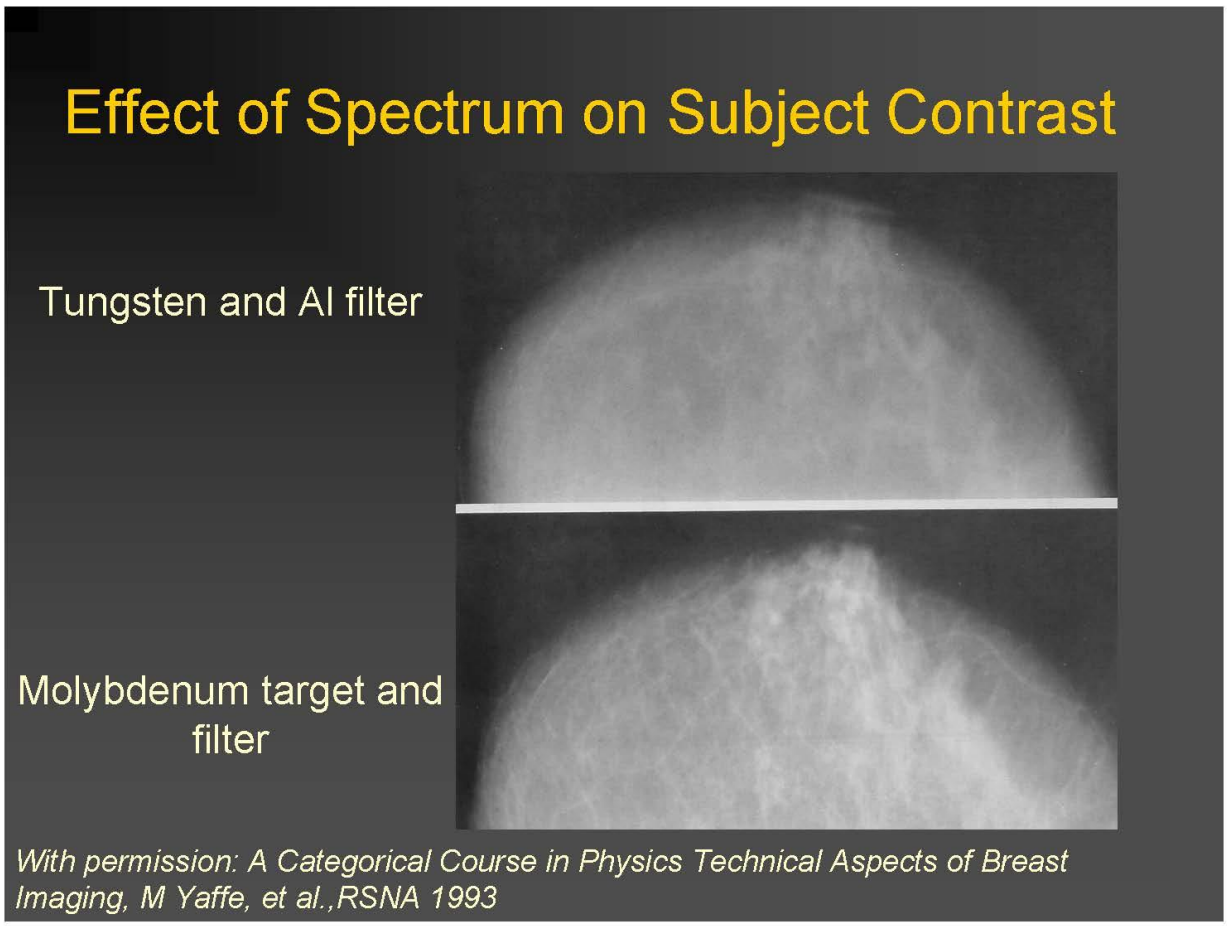
Pathognomonic Signs of Breast Cancer

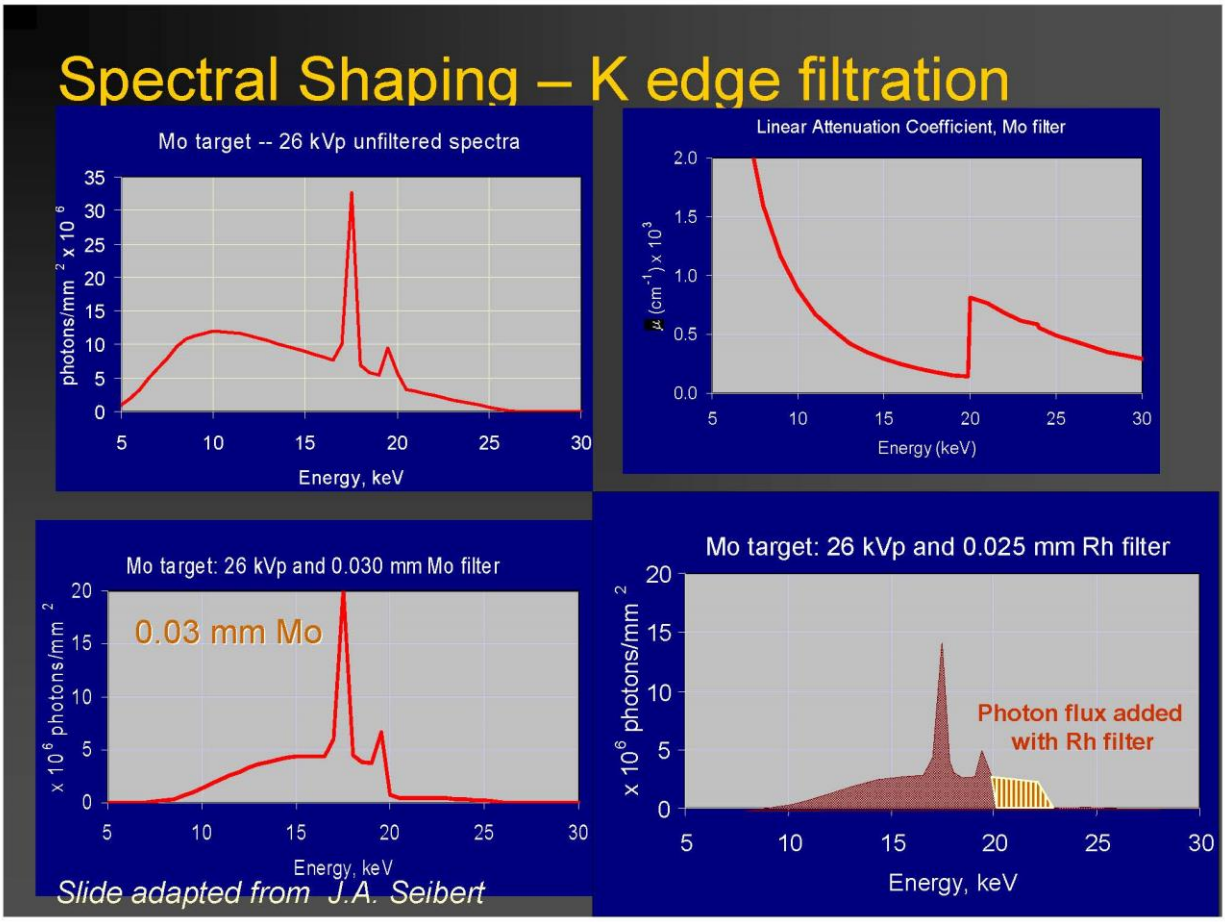
Small Details With Inherent Low Subject Contrast

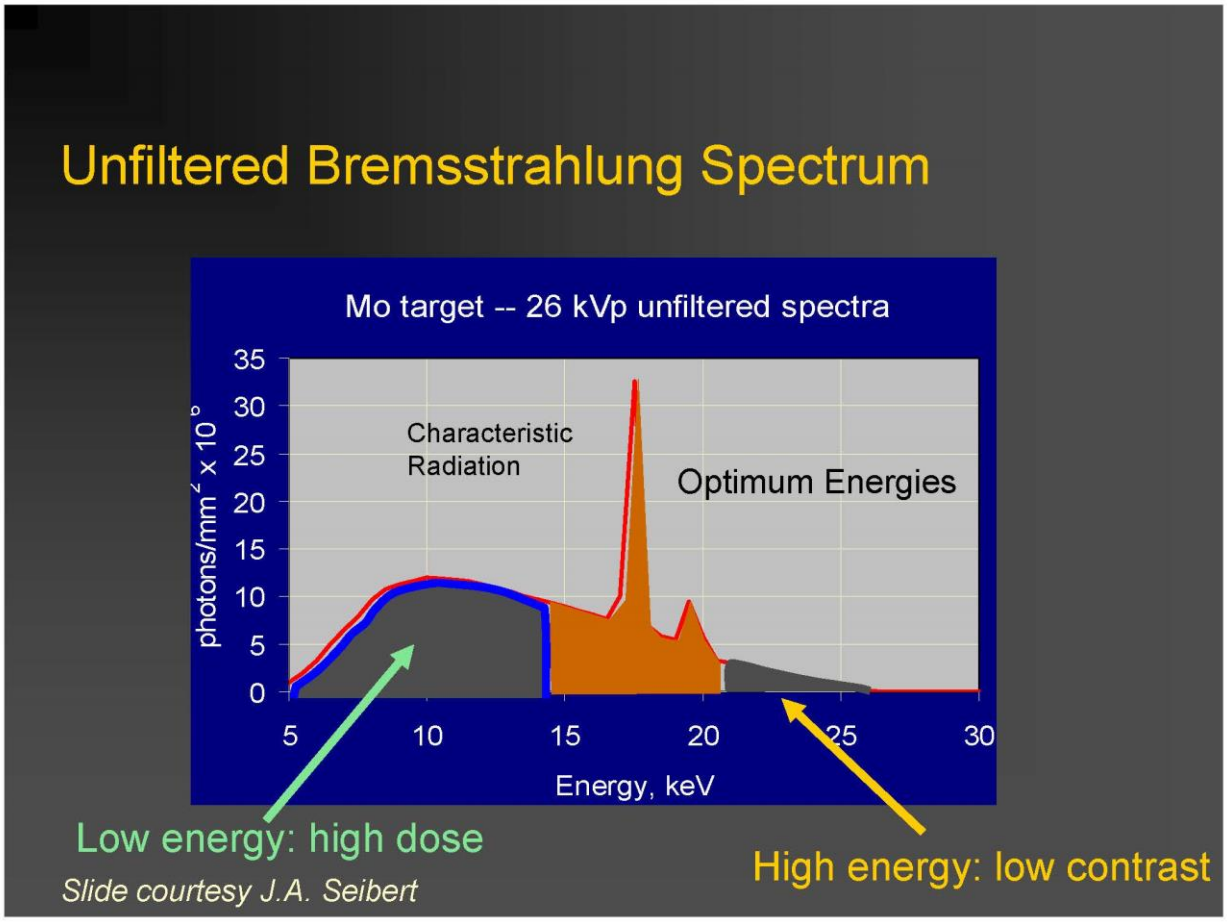
- Masses
  - spiculated
  - shape and margins are important
- Micro-calcifications
  - 100 to 300 microns
  - shape and distribution important
- Others
  - Asymmetric densities
  - Architectural distortions

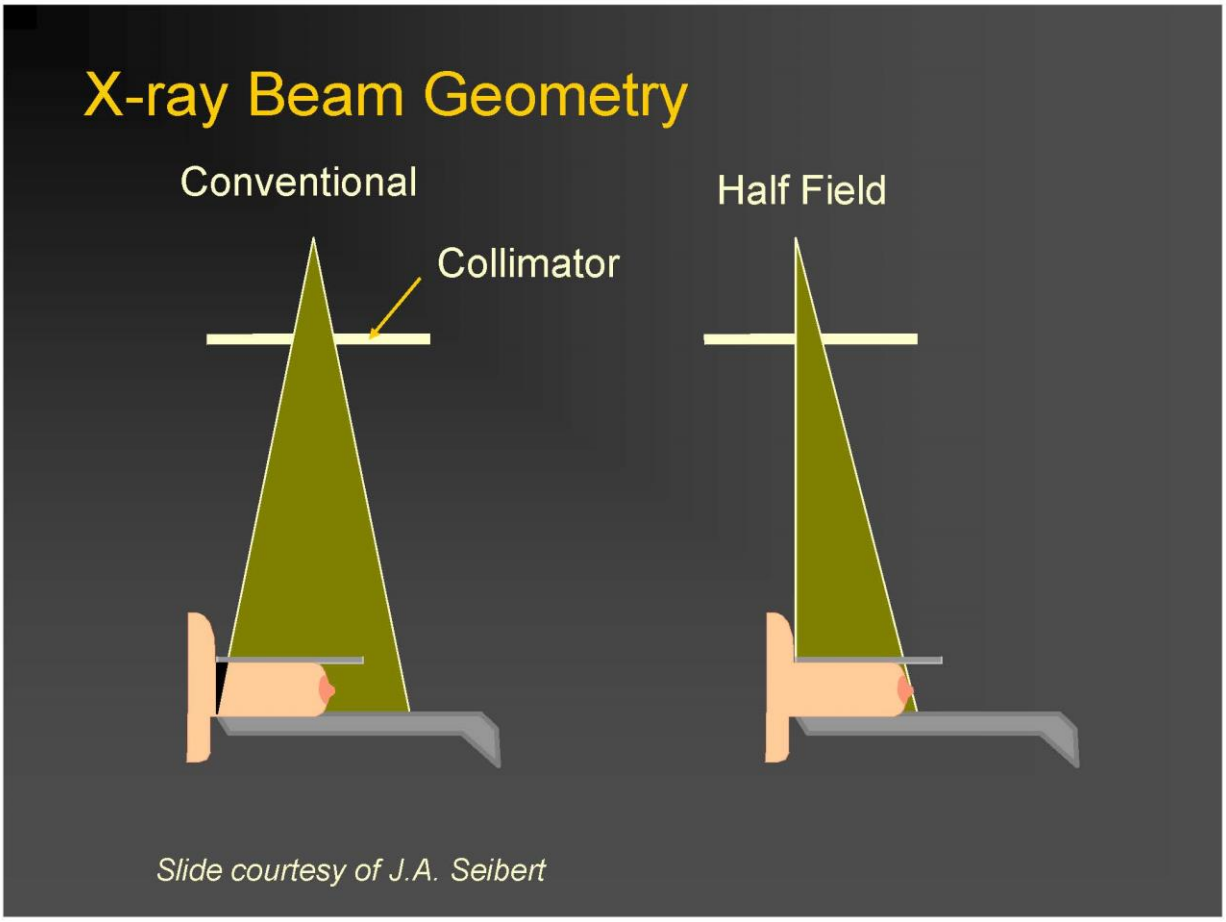


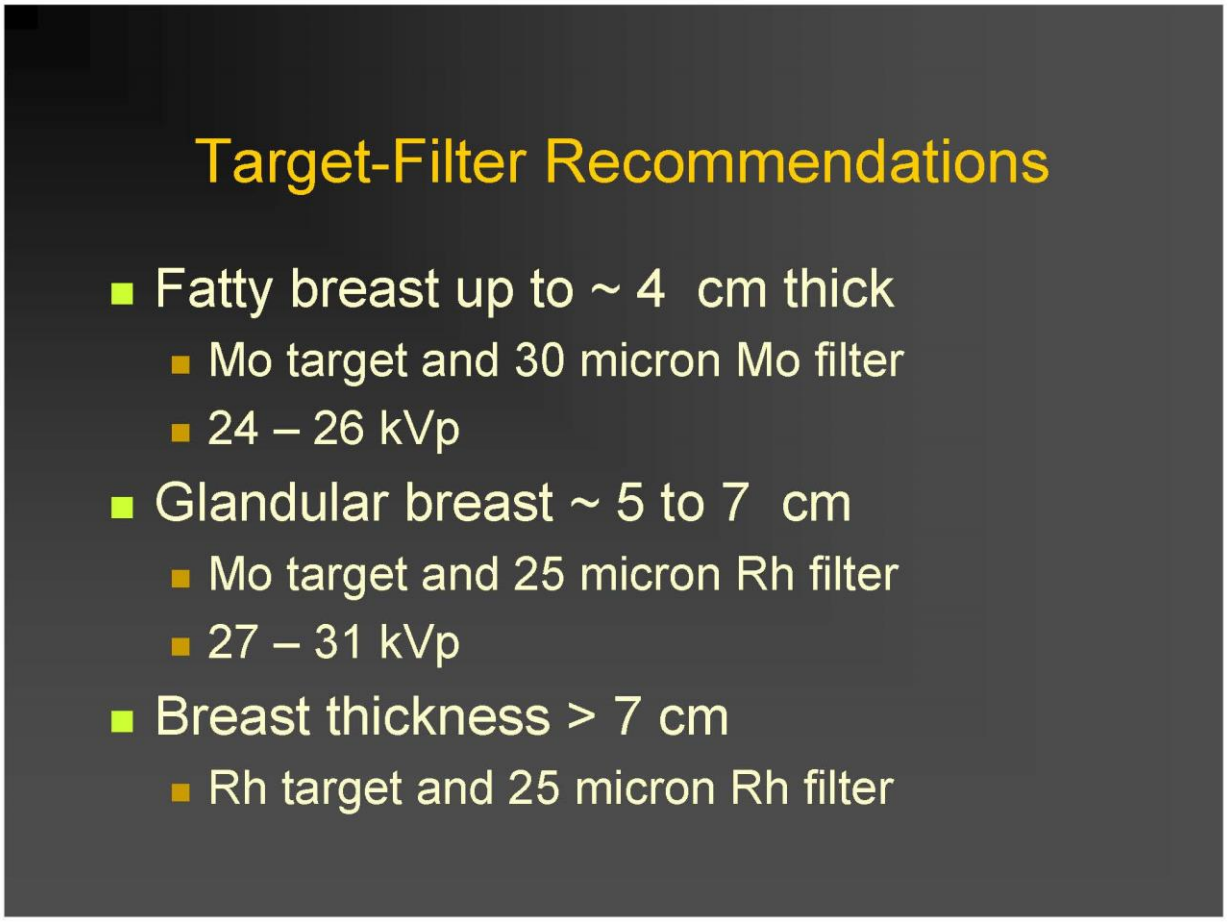

X-Ray Spectrum Shaping
X-ray spectral shaping is needed to enhance visibility of the inherently low contrast pathognomonic signs
Egan
tungsten tube, low kVp, beryllium window tube with minimal aluminum filtration
Gros (CGR)
molybdenum target and molybdenum filter

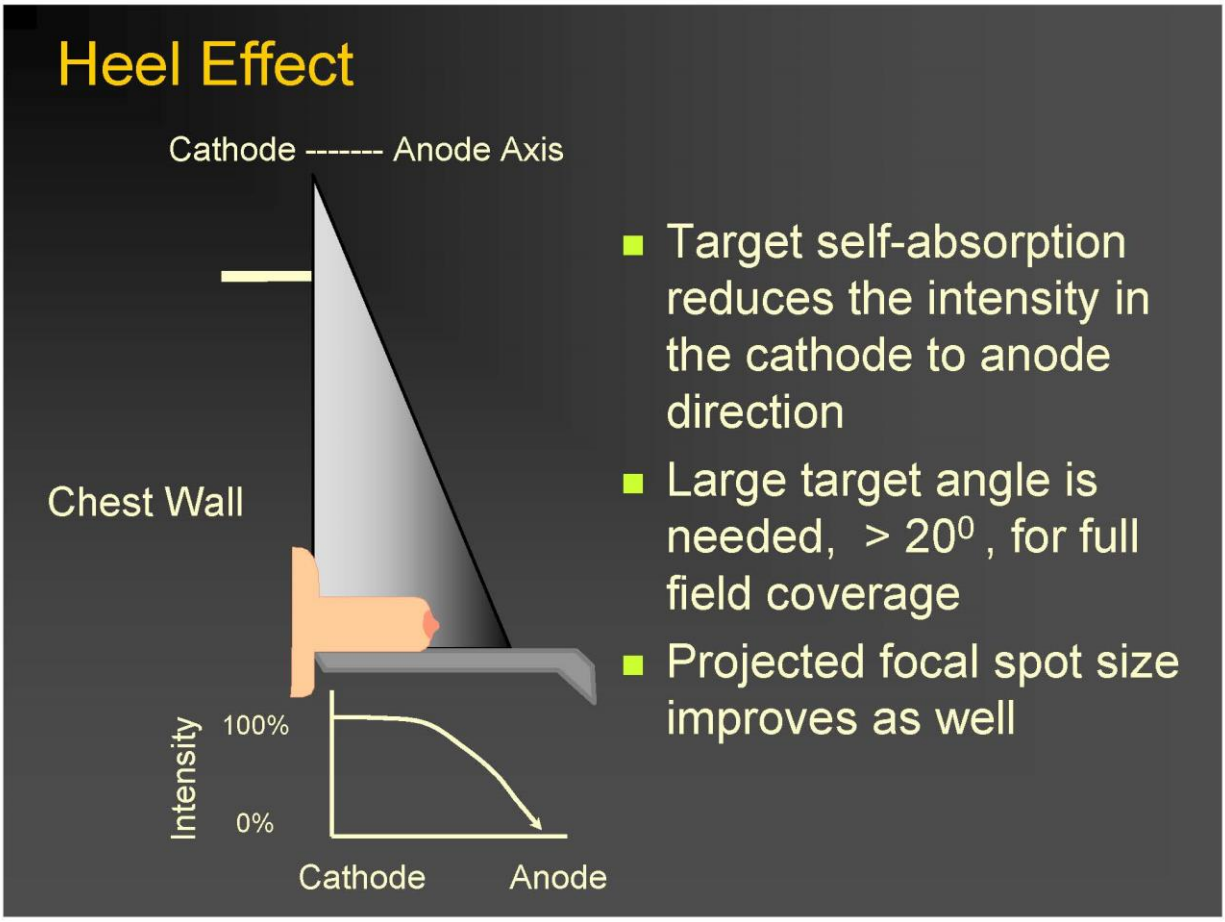

Effect of Spectrum on Subject Contrast
Tungsten and Al filter
Molybdenum target and filter
With permission: A Categorical Course in Physics Technical Aspects of Breast Imaging, M Yaffe, et al.,RSNA 1993


Spectral Shaping – K edge filtration
Mo target -- 26 kVp unfiltered spectra
photons/mm 2 x 10 6
Energy, keV
Linear Attenuation Coefficient, Mo filter
μ (cm-1) x 10 3
Energy (keV)
Mo target: 26 kVp and 0.030 mm Mo filter
0.03 mm Mo
x 10 6 photons/mm 2
Energy, keV
Mo target: 26 kVp and 0.025 mm Rh filter
Photon flux added with Rh filter
x 10 6 photons/mm 2
Energy, keV
Slide adapted from J.A. Seibert


Unfiltered Bremsstrahlung Spectrum
Mo target -- 26 kVp unfiltered spectra
photons/mm² x 10⁶
35
30
25
20
15
10
5
0
Characteristic Radiation
Optimum Energies
5
10
15
20
25
30
Energy, keV
Low energy: high dose
High energy: low contrast
Slide courtesy J.A. Seibert

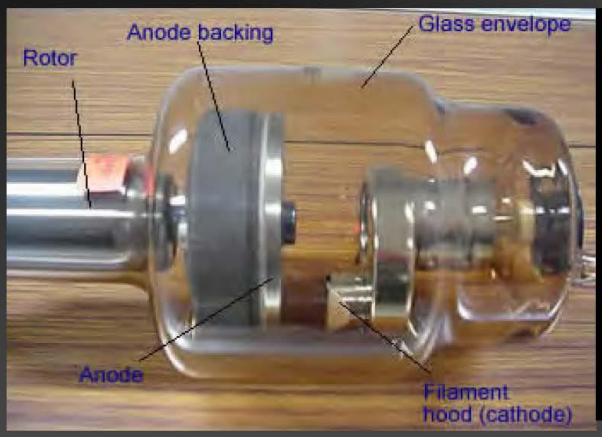

X-ray Beam Geometry
Conventional
Collimator
Half Field
Slide courtesy of J.A. Seibert


Target-Filter Recommendations
Fatty breast up to ~ 4 cm thick
Mo target and 30 micron Mo filter
24 – 26 kVp
Glandular breast ~ 5 to 7 cm
Mo target and 25 micron Rh filter
27 – 31 kVp
Breast thickness > 7 cm
Rh target and 25 micron Rh filter


Heel Effect
Cathode ------- Anode Axis
Chest Wall
Intensity
100%
0%
Cathode
Anode
Target self-absorption reduces the intensity in the cathode to anode direction
Large target angle is needed, > 20⁰ , for full field coverage
Projected focal spot size improves as well

X-Ray Tubes

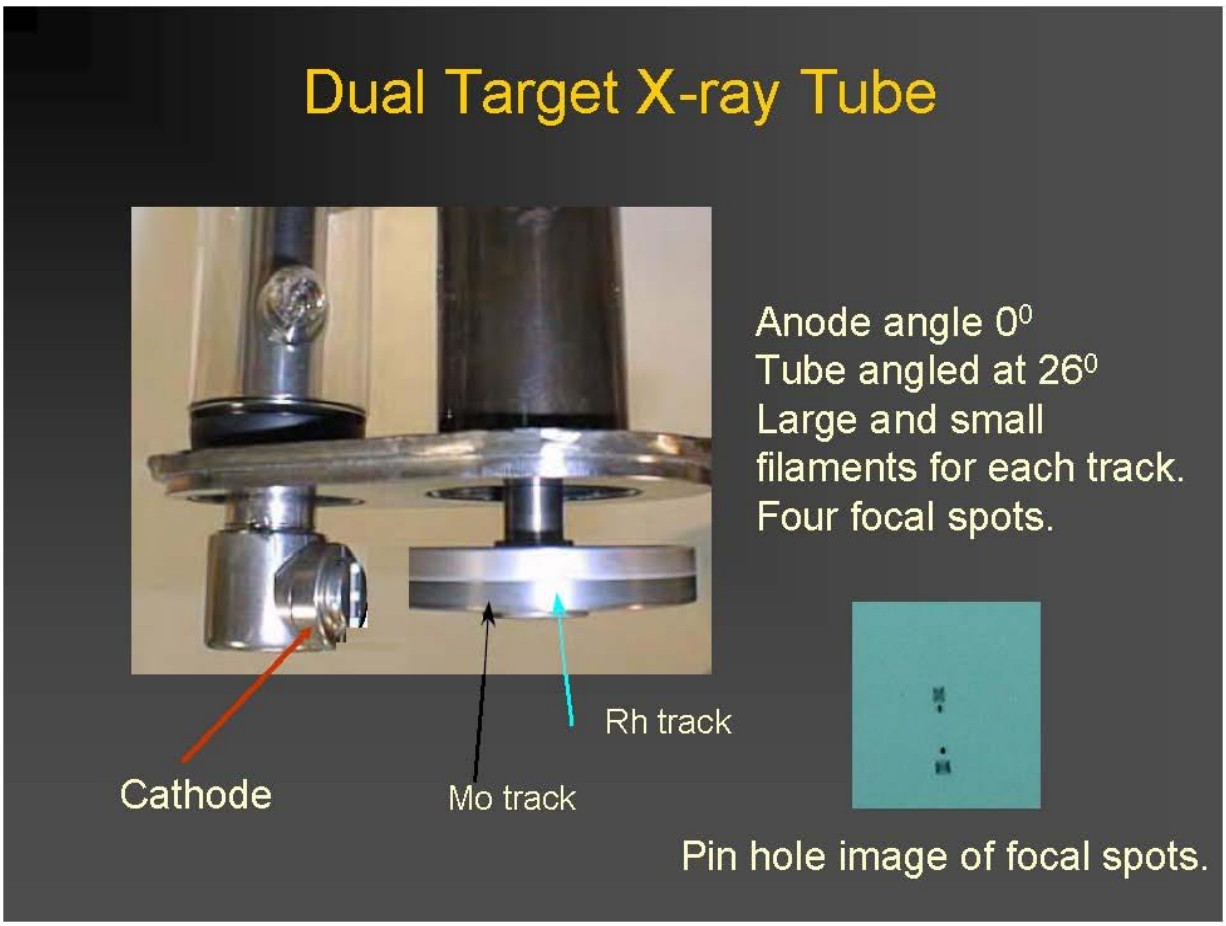
Conventional



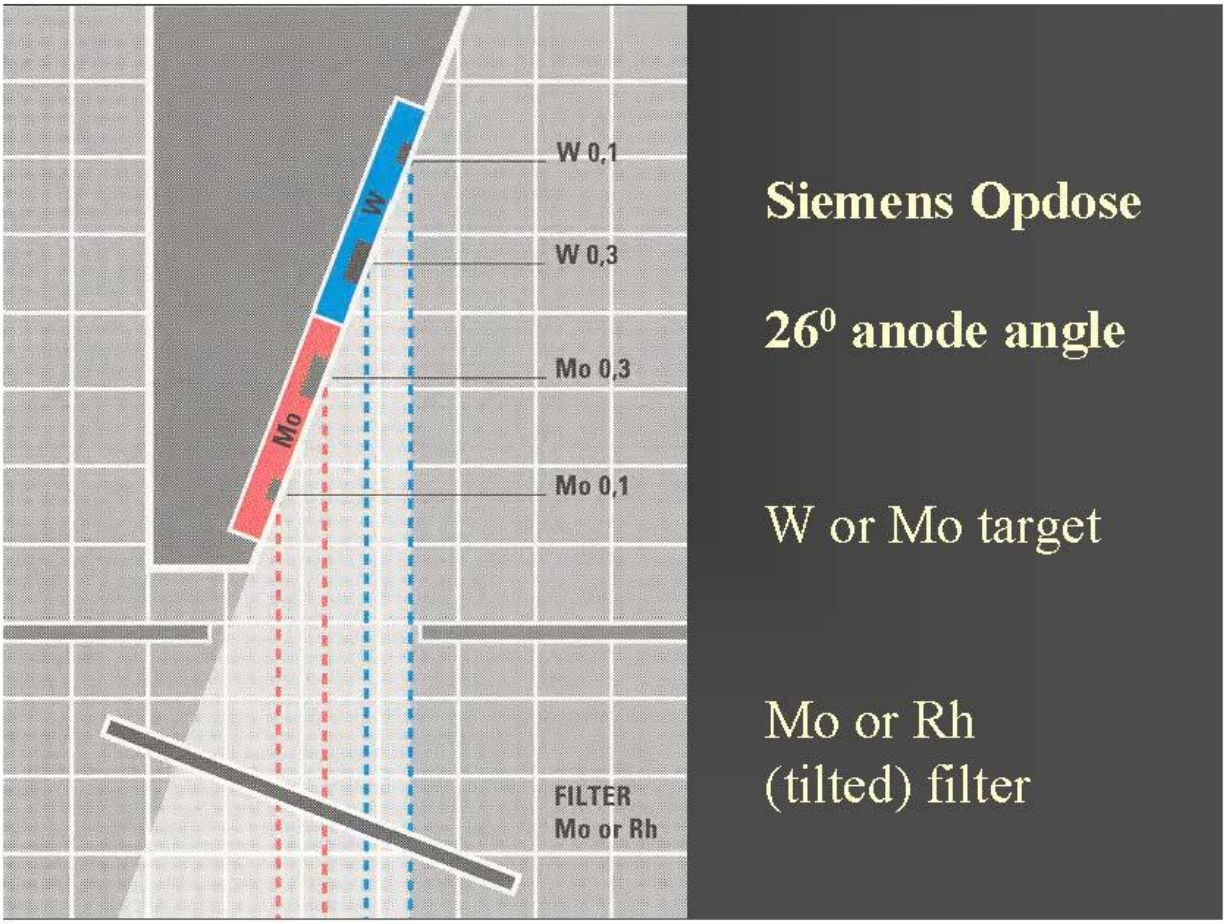
- glass envelope
- tungsten anode
- anode angle ~7^0 to 16^0
- axis of rotation – horizontal
- Al filter for dose reduction

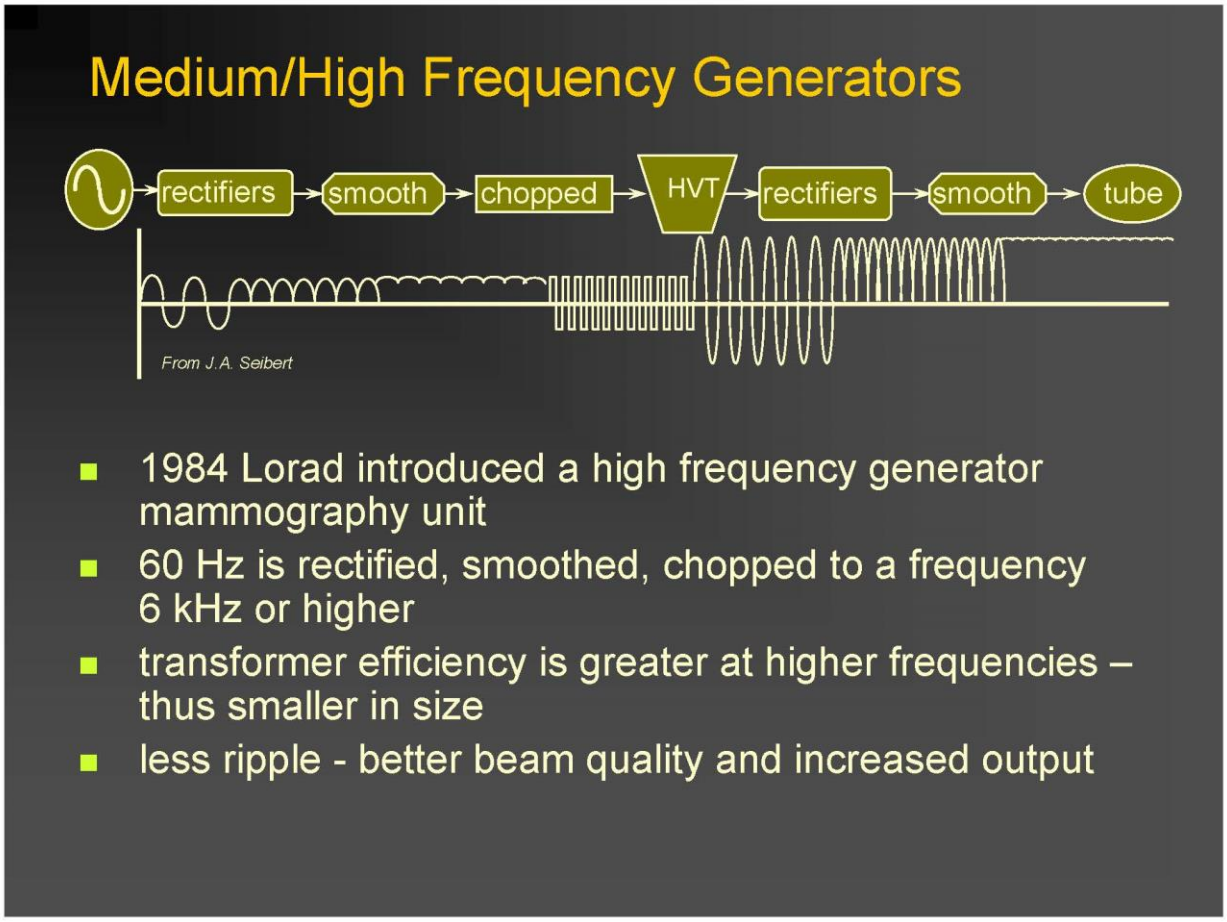
Mammography

- metal tube housing
- grounded Mo, Rh anode
- anode angle 0^0 - tube tilt of 26^0
- axis of rotation ~ vertical
- Mo or Rh filters for spectral shaping


Dual Target X-ray Tube
Anode angle 0⁰
Tube angled at 26⁰
Large and small
filaments for each track.
Four focal spots.
Rh track
Cathode
Mo track
Pin hole image of focal spots.

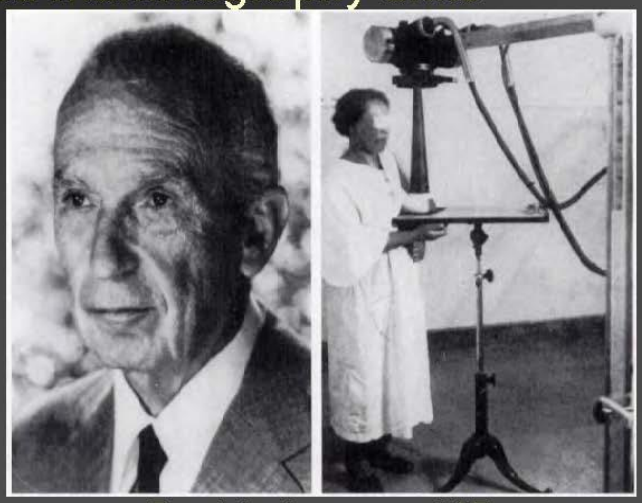

Mammography X-Ray Tube


W 0,1
W 0,3
Mo 0,3
Mo 0,1
W
Mo
FILTER
Mo or Rh
Siemens Opdose
26⁰ anode angle
W or Mo target
Mo or Rh
(tilted) filter

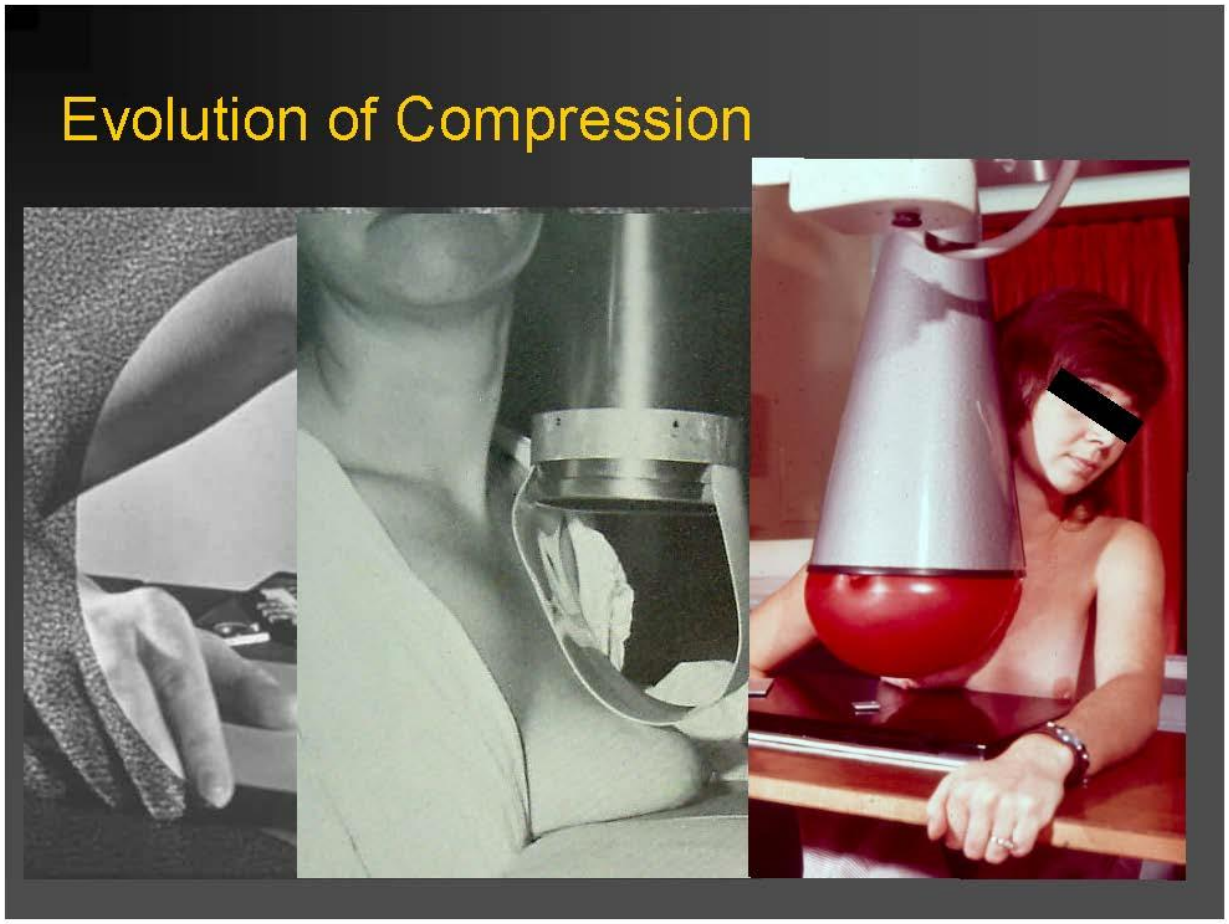

Medium/High Frequency Generators
rectifiers
smooth
chopped
HVT
rectifiers
smooth
tube
From J.A. Seibert
1984 Lorad introduced a high frequency generator mammography unit
60 Hz is rectified, smoothed, chopped to a frequency 6 kHz or higher
transformer efficiency is greater at higher frequencies – thus smaller in size
less ripple - better beam quality and increased output

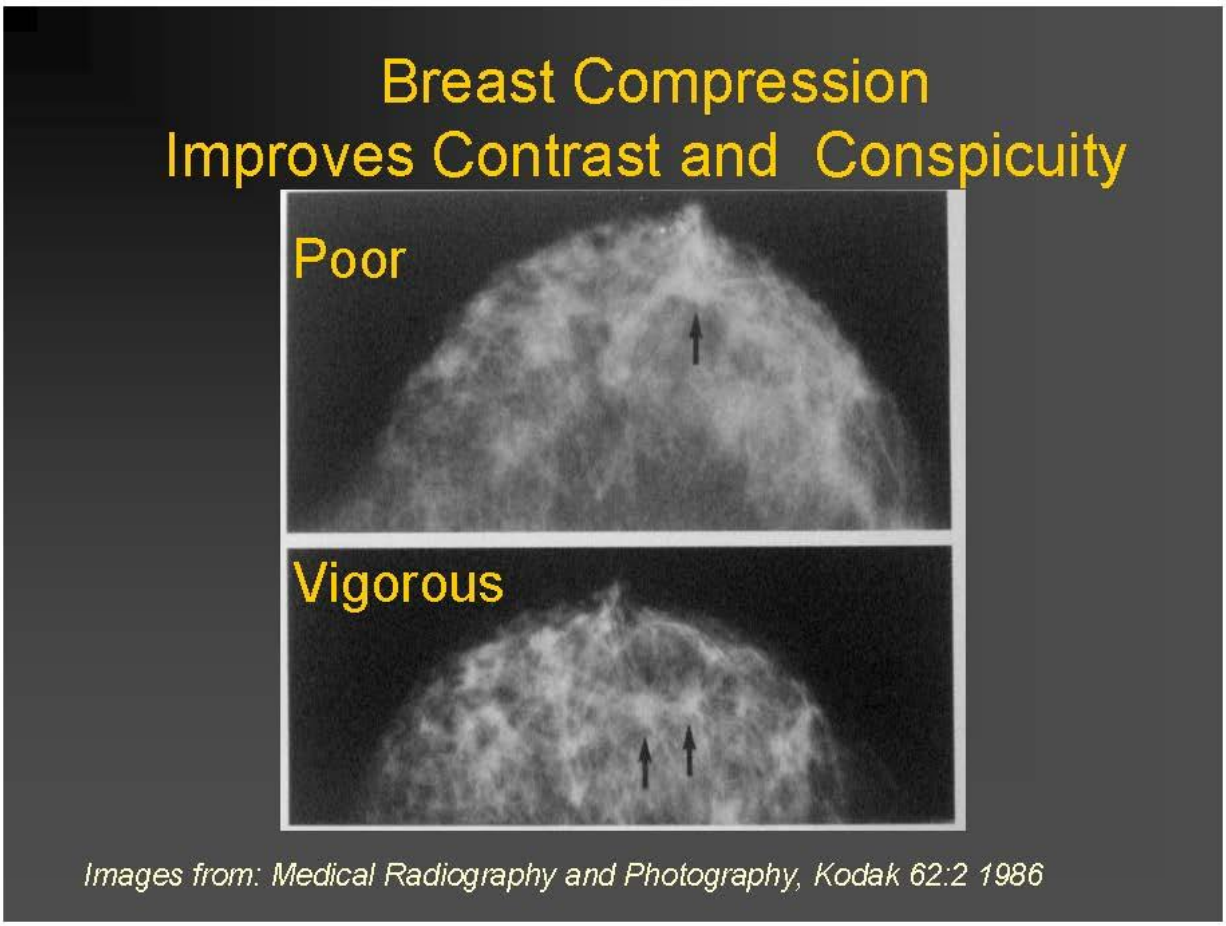
Breast Compression

- 1949 R. Leborgne, Uruguanian radiologist first uses breast compression
- By 1970's compression devices common on dedicated mammography units



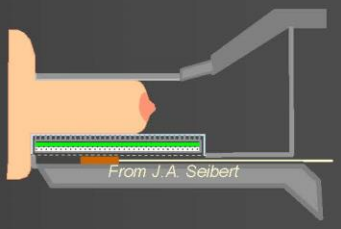
Raul Leborgne, MD


Evolution of Compression

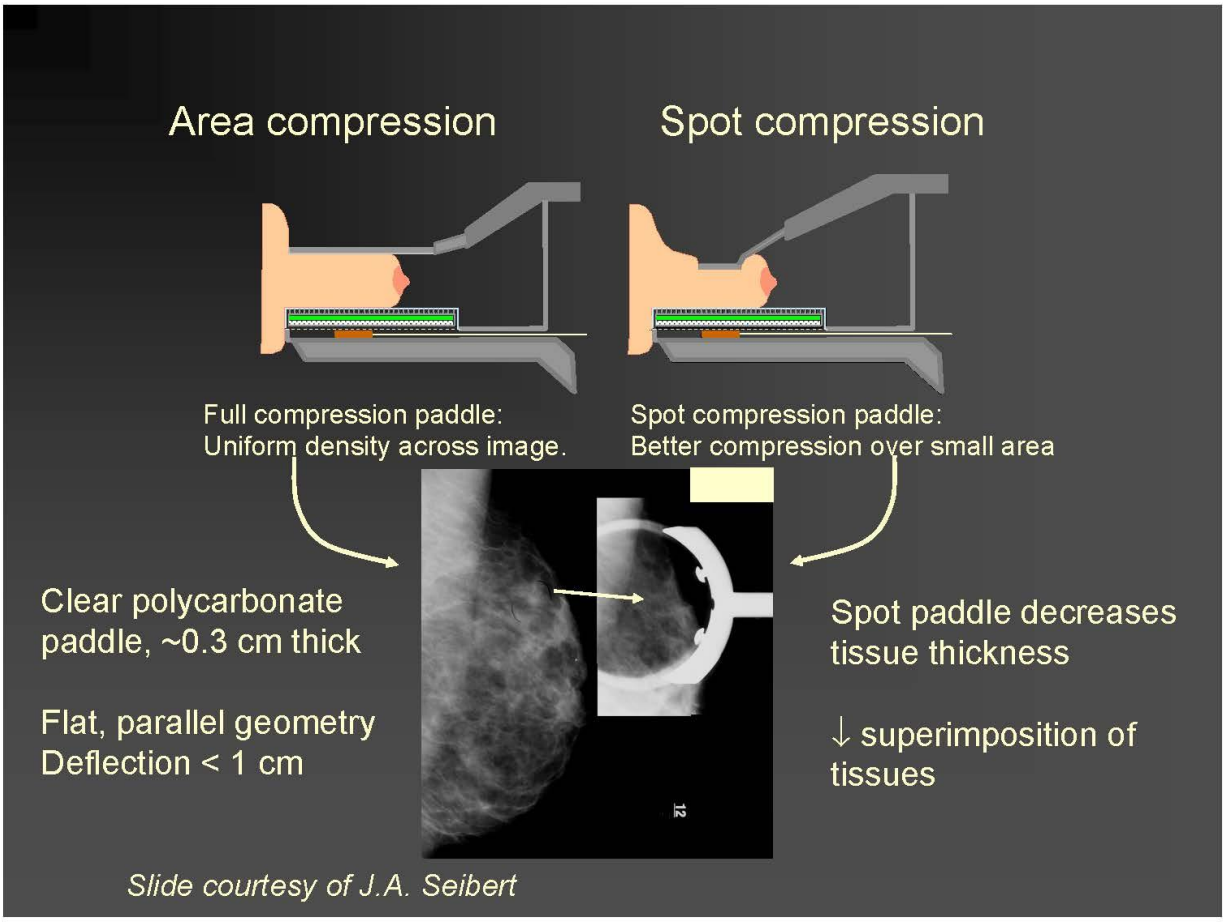

Breast Compression
Improves Contrast and Conspicuity
Poor
Vigorous
Images from: Medical Radiography and Photography, Kodak 62:2 1986

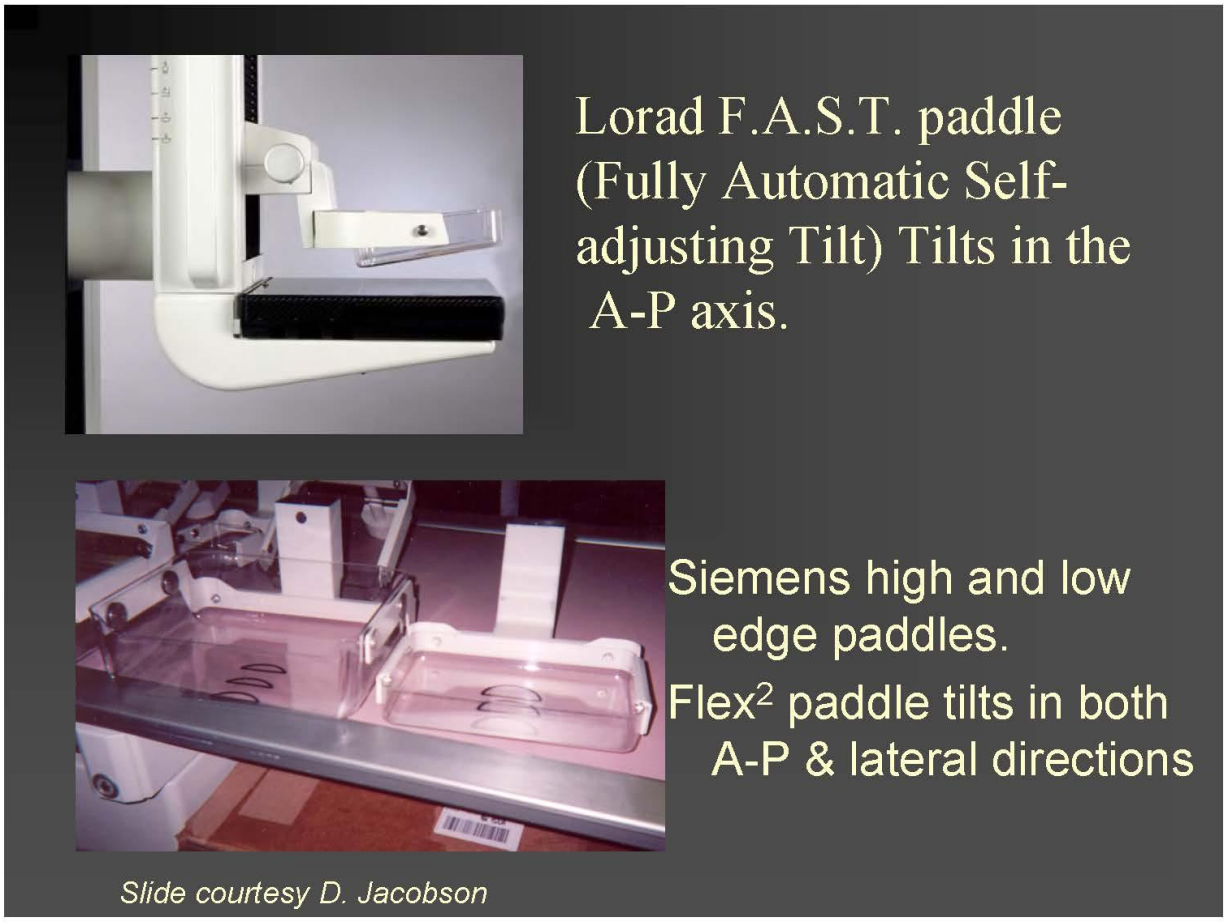
Breast Compression

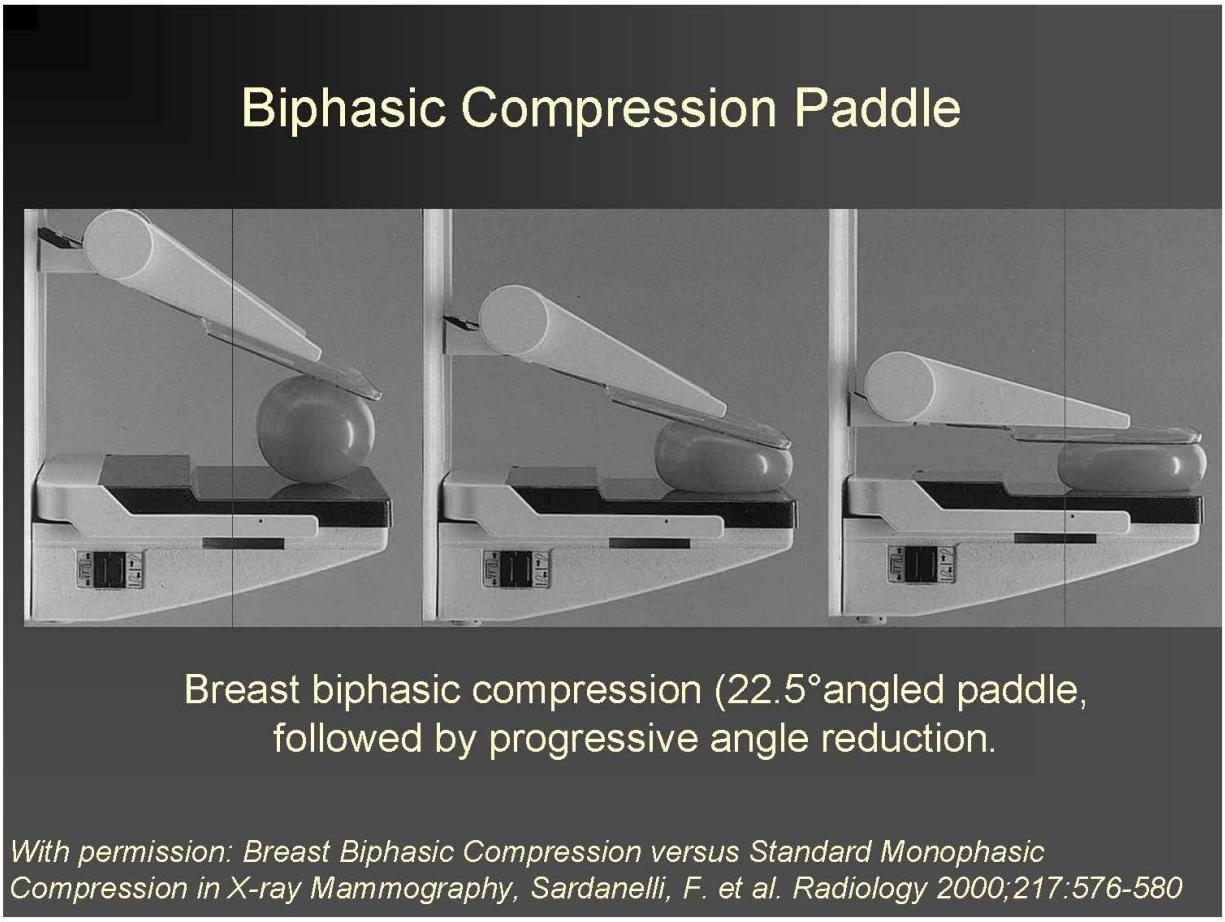
Area compression

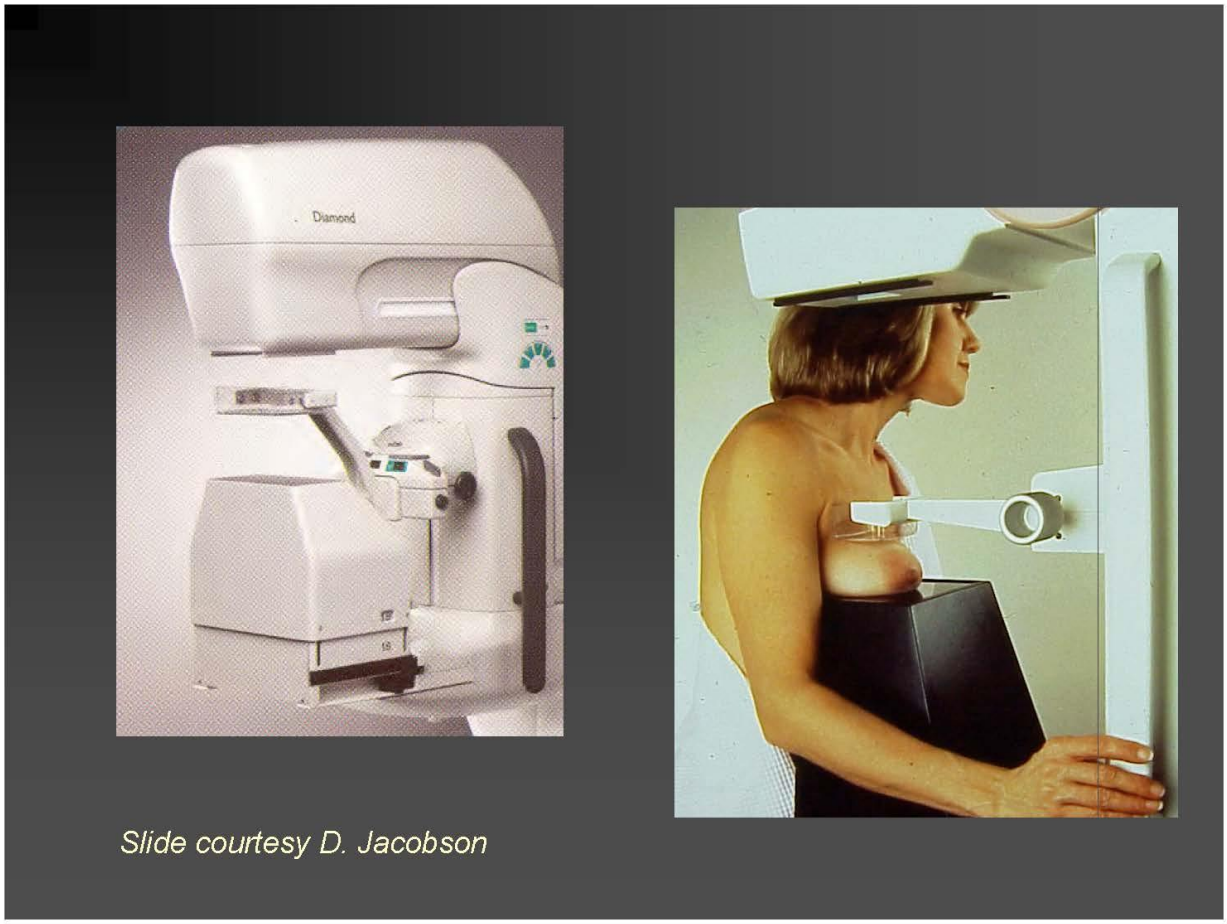


- Reduces breast thickness
  - lowers radiation dose
  - spreads breast tissues apart
  - produces a more uniform thickness
    - allows use of narrow latitude, high contrast film
- Reduces motion and geometric unsharpness
- Reduces x-ray scatter and beam hardening, thus improving contrast


Area compression
Spot compression
Full compression paddle:
Uniform density across image.
Spot compression paddle:
Better compression over small area
Clear polycarbonate paddle, ~0.3 cm thick
Flat, parallel geometry
Deflection < 1 cm
Spot paddle decreases tissue thickness
↓ superimposition of tissues
Slide courtesy of J.A. Seibert

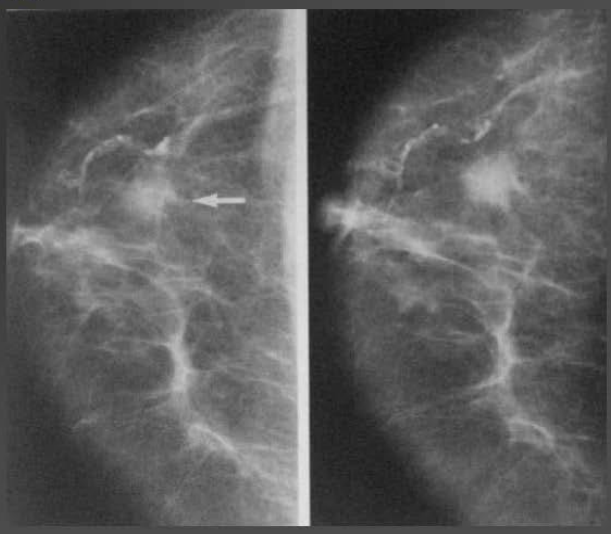

Lorad F.A.S.T. paddle (Fully Automatic Self-adjusting Tilt) Tilts in the A-P axis.
Siemens high and low edge paddles.
Flex2 paddle tilts in both A-P & lateral directions
Slide courtesy D. Jacobson


Biphasic Compression Paddle
Breast biphasic compression (22.5°angled paddle, followed by progressive angle reduction.
With permission: Breast Biphasic Compression versus Standard Monophasic Compression in X-ray Mammography, Sardanelli, F. et al. Radiology 2000;217:576-580


Diamond
Slide courtesy D. Jacobson

Anti-scatter Grids

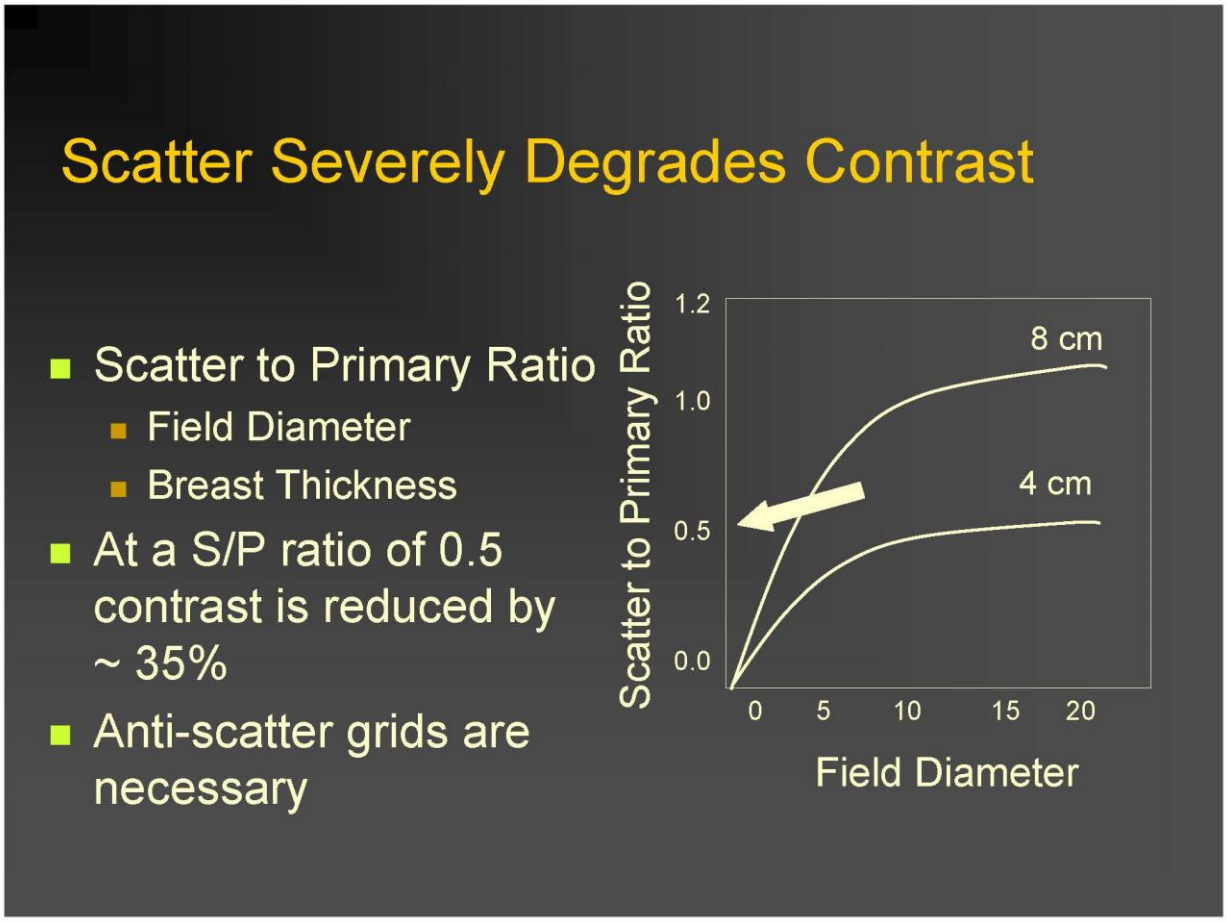
- 1978
  - Philips introduces the Diagnost-U with a moving grid
- 1984
  - Leibel-Flarsheim introduces fine-line stationary grid



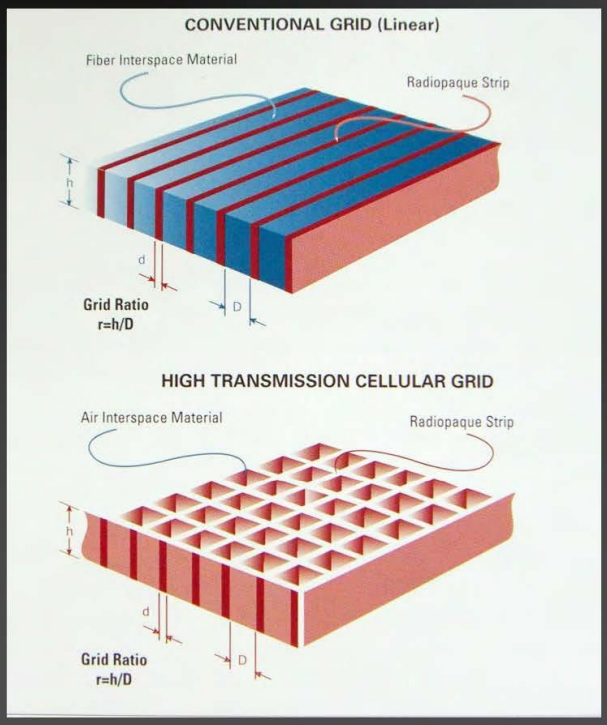
Images from: Medical Radiography and Photography, Kodak 62:2 1986

No grid
26 kVp

With grid
28 kVp

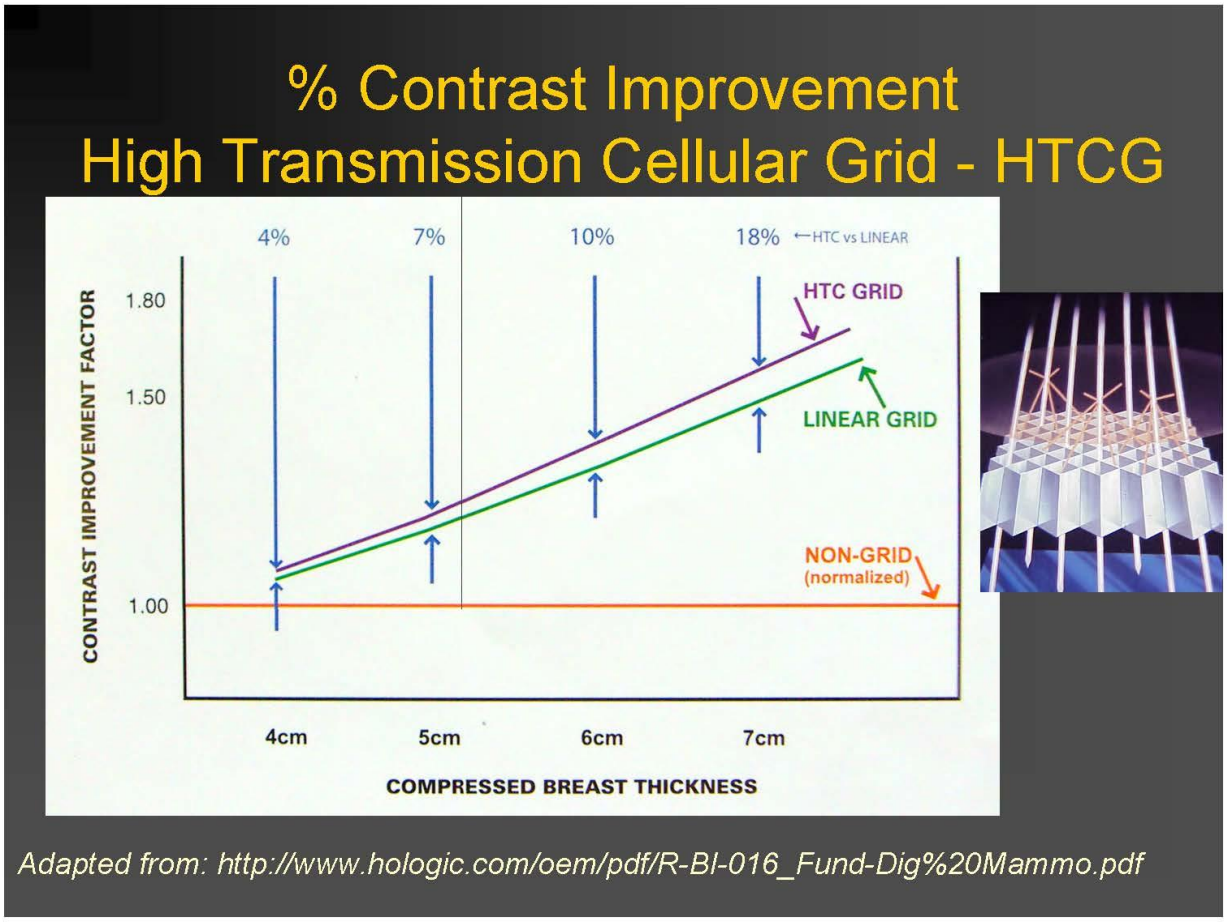

Scatter Severely Degrades Contrast
Scatter to Primary Ratio
Field Diameter
Breast Thickness
At a S/P ratio of 0.5 contrast is reduced by ~ 35%
Anti-scatter grids are necessary
Scatter to Primary Ratio
1.2
1.0
0.5
0.0
8 cm
4 cm
0
5
10
15
20
Field Diameter

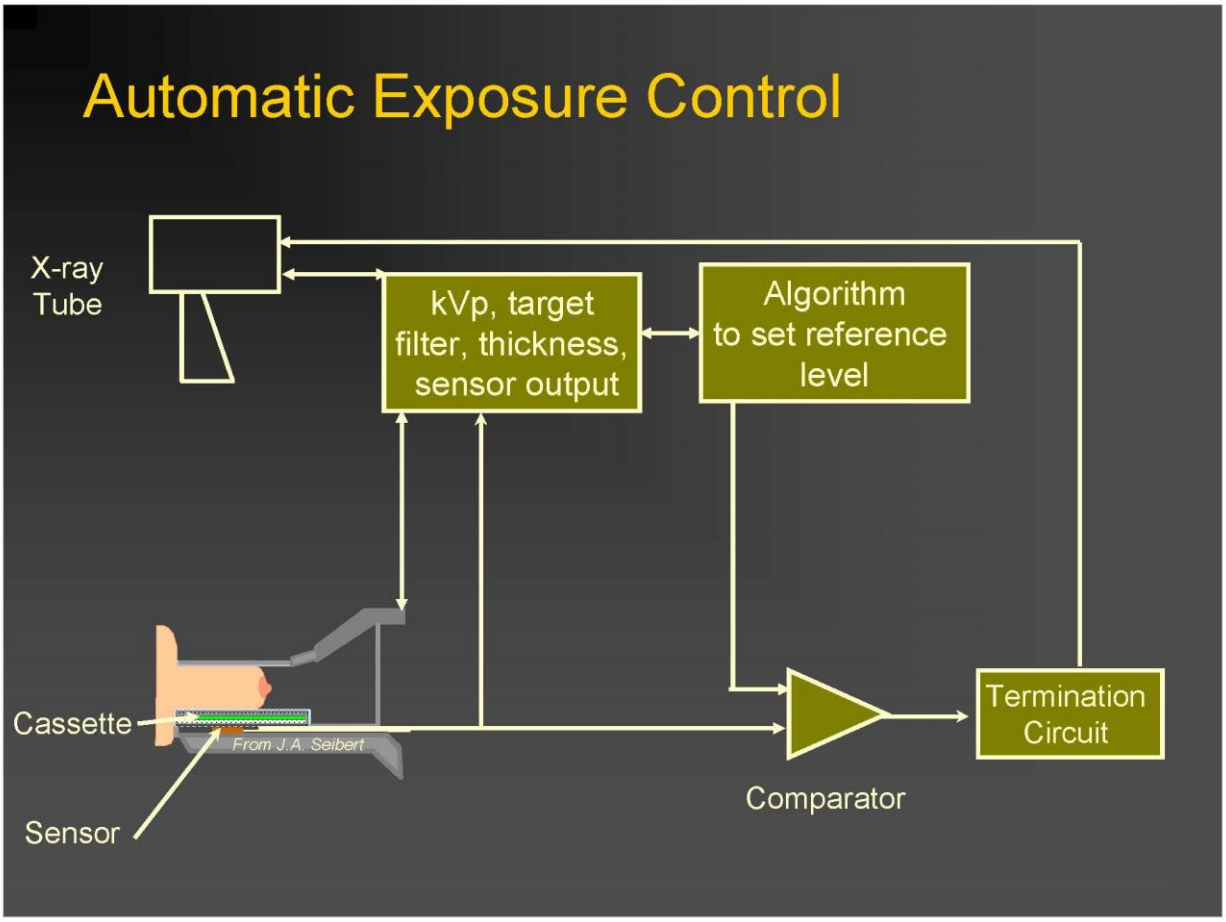
Grids



- Linear grid
  - Ratio: 5:1
  - Frequency > 30 l/cm
  - Wood, paper or carbon fiber inter-space material
  - Moved ~20 lines for blurring
- Cellular grid
  - 15 cells / cm
  - Air inter-space
  - Moved multiple of hole spacing

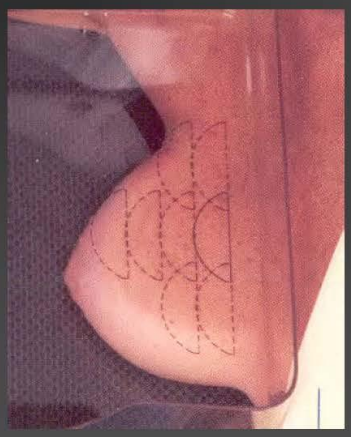
Figure from: http://www.hologic.com/oem/pdf/R-BI-016_Fund-Dig%20Mammo.pdf


% Contrast Improvement
High Transmission Cellular Grid - HTCG
4%
7%
10%
18%
←HTC vs LINEAR
HTC GRID
LINEAR GRID
NON-GRID
(normalized)
CONTRAST IMPROVEMENT FACTOR
1.80
1.50
1.00
4cm
5cm
6cm
7cm
COMPRESSED BREAST THICKNESS
Adapted from: http://www.hologic.com/oem/pdf/R-BI-016_Fund-Dig%20Mammo.pdf

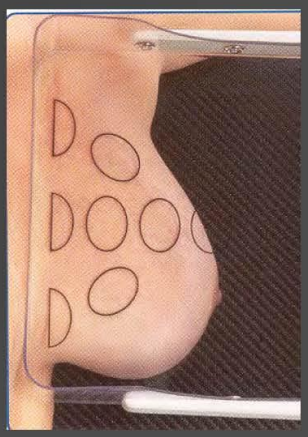

Automatic Exposure Control
X-ray Tube
kVp, target filter, thickness, sensor output
Algorithm to set reference level
Termination Circuit
Cassette
From J.A. Seibert
Sensor
Comparator

Automatic Exposure Control

- AEC sensor is located *underneath* the cassette
  - typical screen exposure is 5 to 10 mR
  - variable sensor position
  - should be under densest tissue



GE Instrumentarium
Vector Point



GE Instrumentarium
Diamond Autopoint

Automatic Exposure Control

- AEC sensor is located *underneath* the cassette
  - typical screen exposure is 5 to 10 mR
  - variable sensor position
  - should be under densest tissue
  - integrated signal is used to terminate the exposure

GE Instrumentarium Diamond Autopoint

AEC Modes of Operation

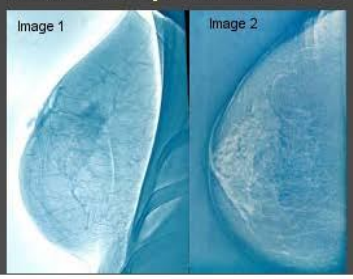
- Auto Time
  - kVp, target/filter chosen by operator
- Auto kVp
  - kVp chosen on basis of breast thickness
- Full Automatic
  - kVp, target/filter chosen by unit
- Siemens – Opdose
  - Breast thickness used to suggest kVp and target/ filter combination
- GE Instrumentarium
  - kVp adjusted during exposure to achieve exposure time of ~2 seconds
- GE DMR
  - Attenuation (100 ms) and breast thickness are used to select kVp and target/filter combination
  - Three algorithms – STD, DOSE and CNT

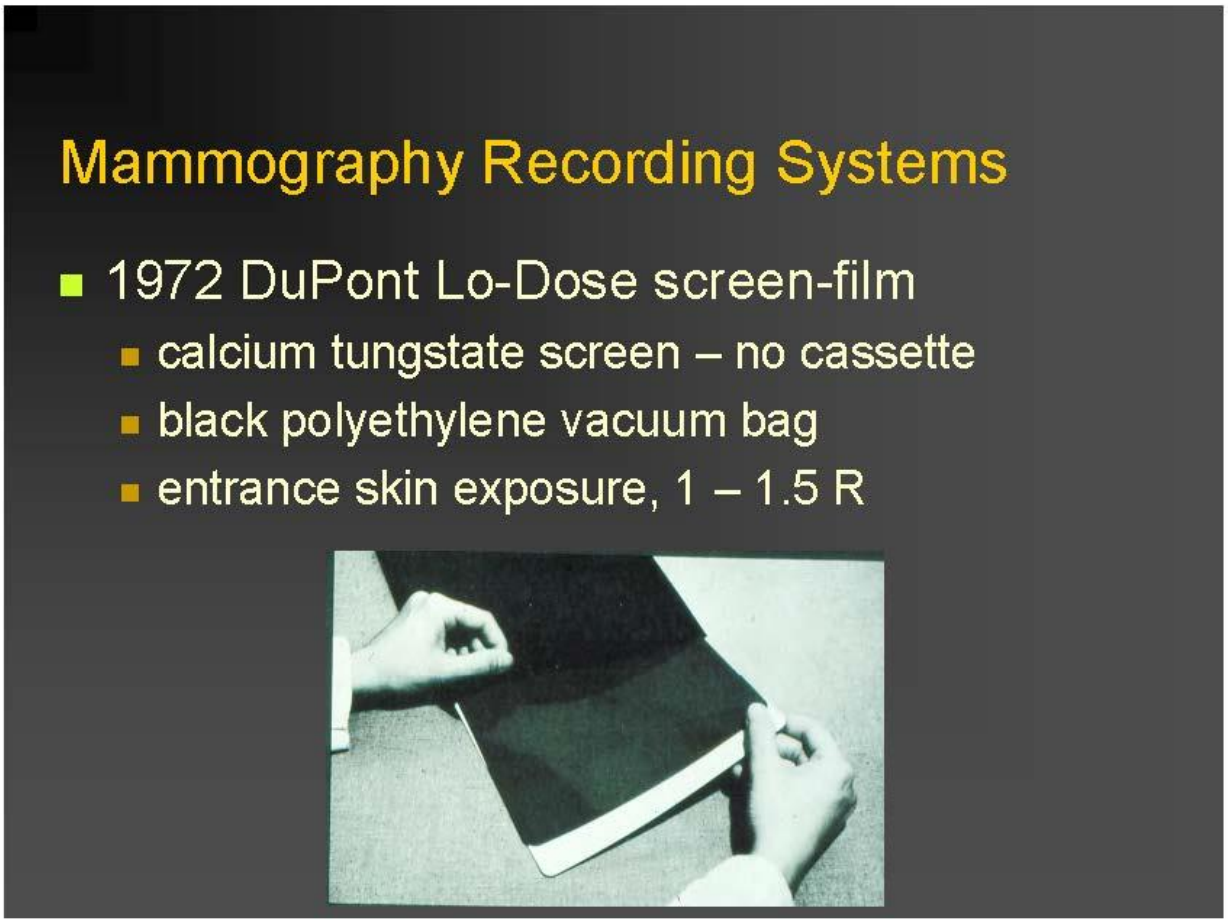
Mammographic Recording Systems

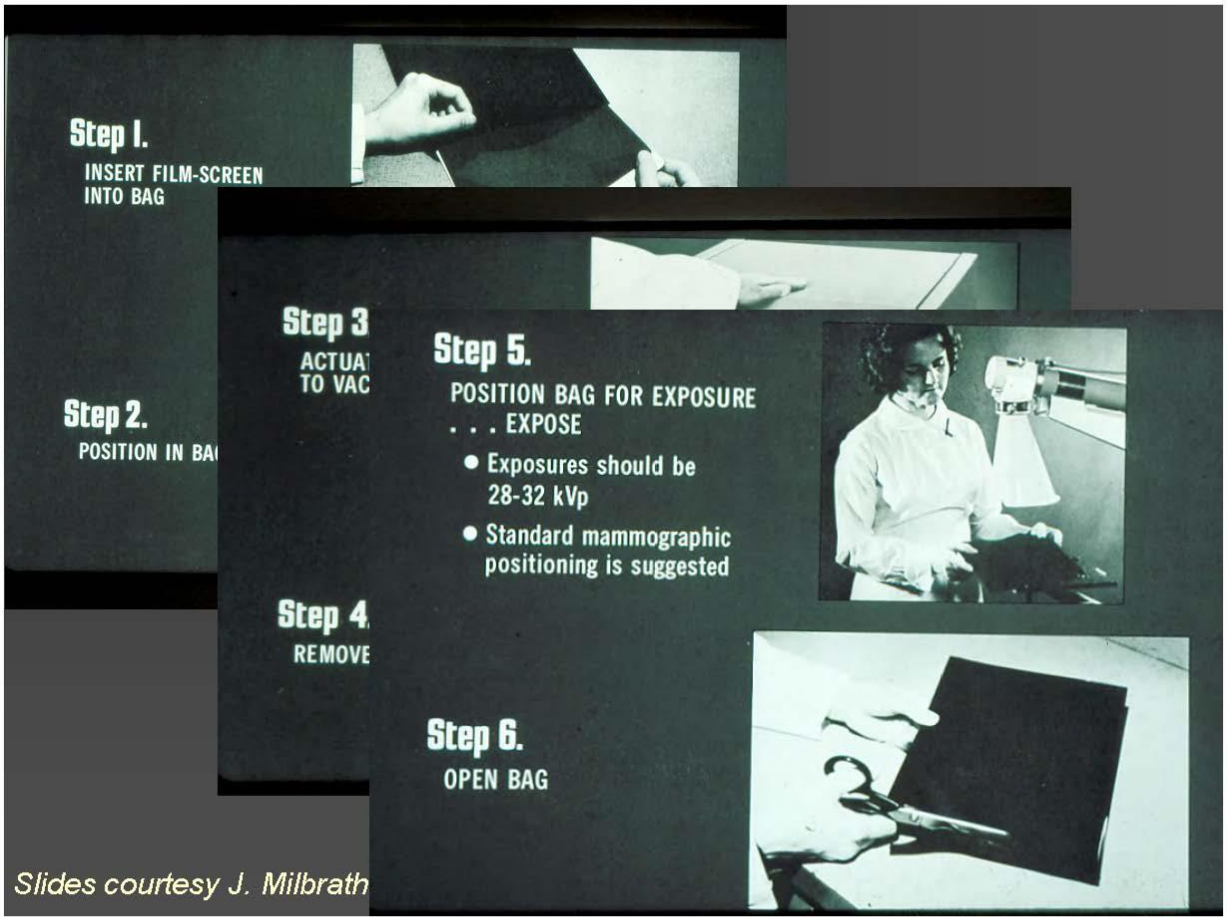
- 1960's non-screen industrial film
  - hand processing – 5 minutes
- 1970 Kodak RP/M non-screen film
  - 90 second processing
  - entrance skin exposure, 3 – 10 R

Mammography Recording Systems

- 1950's non-screen industrial film
- 1970 Kodak RP/M non-screen film
  - 90 second processing
  - entrance skin exposure, 3 – 10 R
- 1971 Xeroradiography
  - blue powder
  - entrance skin exposure, 2 – 4 R




Mammography Recording Systems
1972 DuPont Lo-Dose screen-film
calcium tungstate screen – no cassette
black polyethylene vacuum bag
entrance skin exposure, 1 – 1.5 R

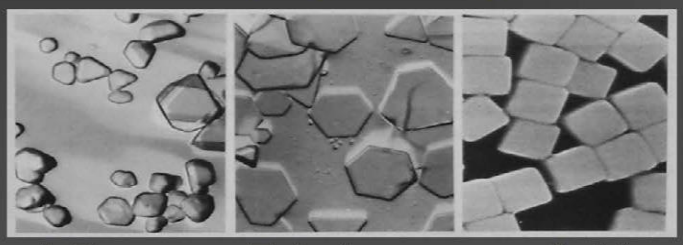

Step I.
INSERT FILM-SCREEN
INTO BAG
Step 2.
POSITION IN BA
Step 3
ACTUA
TO VAC
Step 4
REMOVE
Step 5.
POSITION BAG FOR EXPOSURE
. . . EXPOSE
• Exposures should be 28-32 kVp
• Standard mammographic positioning is suggested
Step 6.
OPEN BAG
Slides courtesy J. Milbrath

Mammography Recording Systems

- 1972 DuPont Lo-Dose screen-film
  - calcium tungstate screen
  - black polyethylene vacuum bag
  - entrance skin exposure, 1 – 1.5 R
- 1976 DuPont Lo-Dose II
  - rare-earth screen, cassette
- 1976 Kodak MinR
  - rare-earth screen, cassette

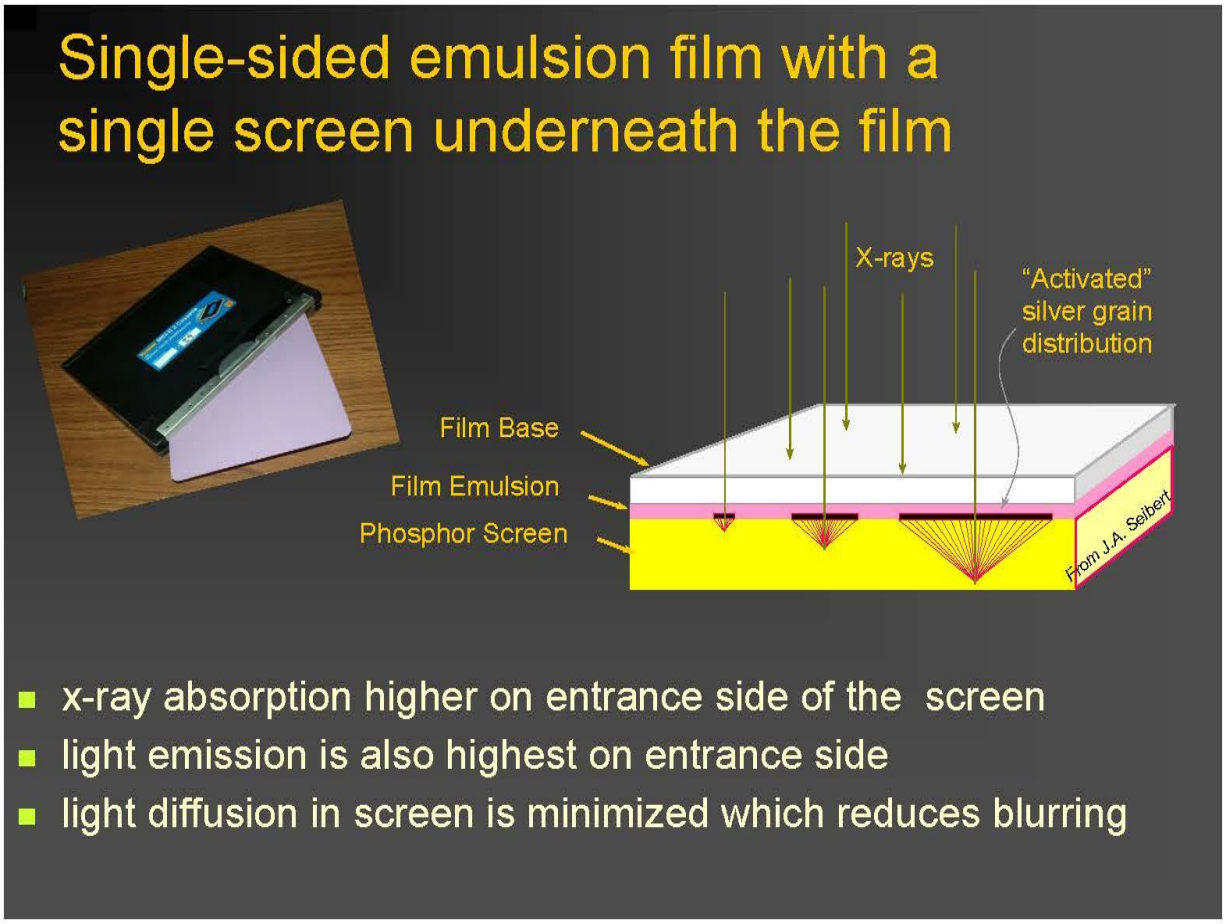
Mammography Recording Systems

- 1983 Kodak Min- R screen-film system
  - gadolinium oxysulfide with orthochromatic film
    - (other rare earth phosphors developed)
  - significant reduction in dose compared to non-screen film
  - current films employ cubic grains



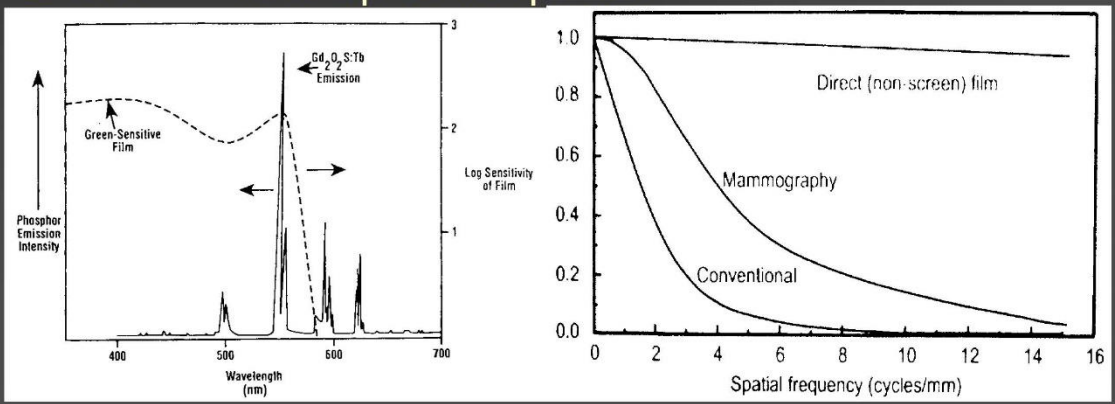
3-D Tabular Cubic

With permission: Medical Physics Publishing, "The Basics of Film Processing in Medical Imaging" by Art Haus and Susan Jaskulski


Single-sided emulsion film with a single screen underneath the film
X-rays
"Activated" silver grain distribution
Film Base
Film Emulsion
Phosphor Screen
From J.A. Seibert
x-ray absorption higher on entrance side of the screen
light emission is also highest on entrance side
light diffusion in screen is minimized which reduces blurring

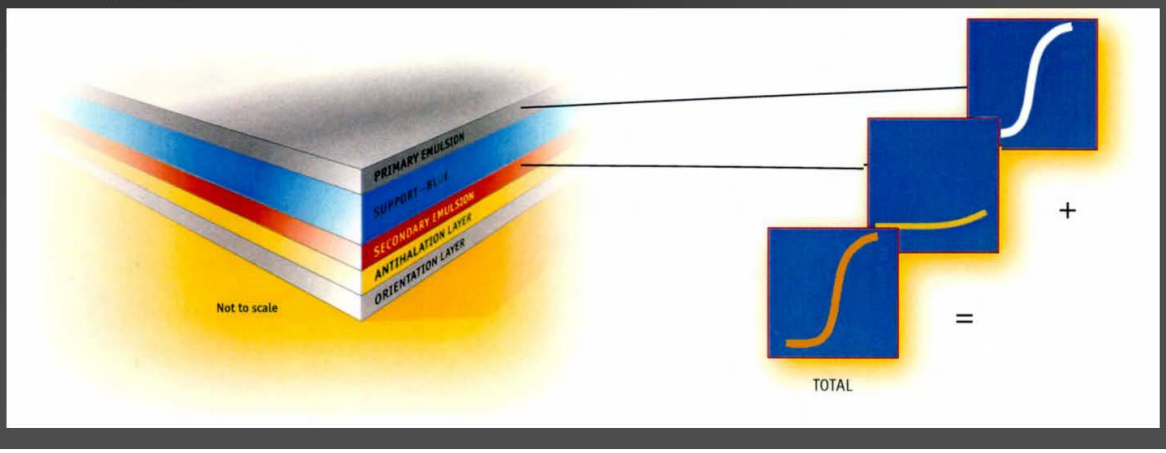
Typical Screen: Gd_2O_2S:Tb

- main emission at 545 nm
  - film spectral sensitivity is matched
- conversion efficiency ~ 15 %
- x-ray absorption of 40 to 60%
- MTF ~ 10% up to 15 lp/mm



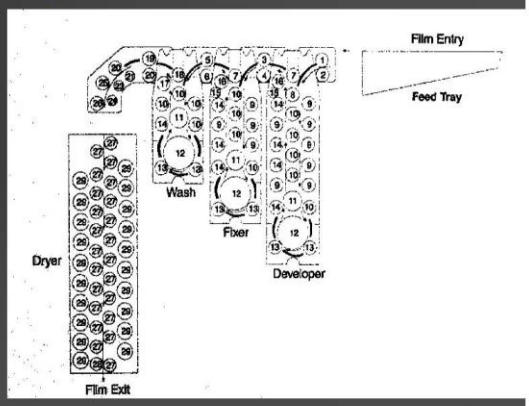
Mammography Recording Systems

- 2003 Kodak Min-R EV
  - Dual emulsion film used with a single screen
  - Asymmetric emulsion design optimizes image quality from toe to shoulder of the sensitometric curve



Film Exposure and Processing

- Latent image formation
  - Light converts AgBr complex into silver ion + electron, creates a sensitivity speck
- Processing (four steps)
  - Developer
    - Chemical amplification ~ 5×10^9
  - Fixer stops development
  - Washing
  - Drying

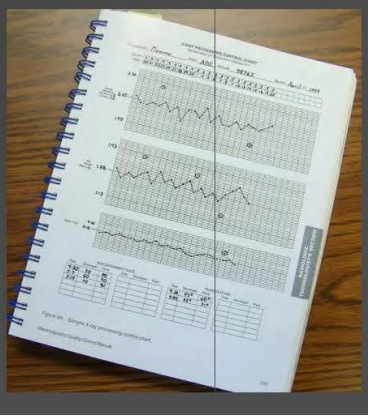


For details see: "The Basics of Film Processing in Medical Imaging" by Art Haus and Susan Jaskulski

Technologist Daily Processor Control

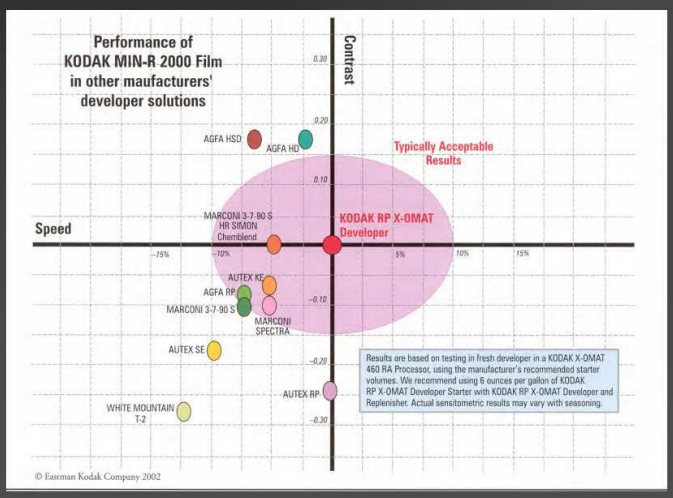
MQSA requires a processor performance test on each day that examinations are performed before any clinical films are processed

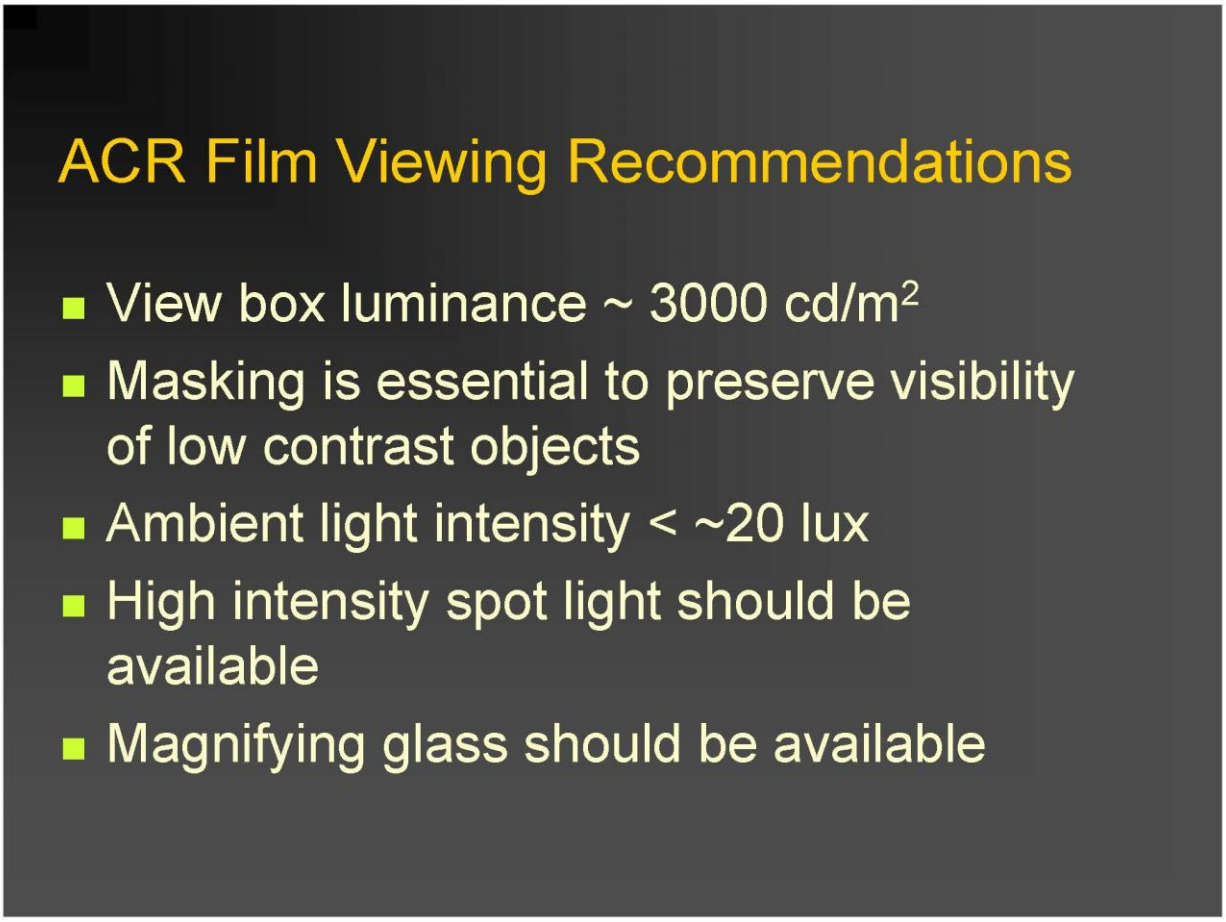
- Density Difference
  - +/- 0.15 DU
- Mid-density
  - +/- 0.15 DU
- Base + Fog
  - +/- 0.03 DU

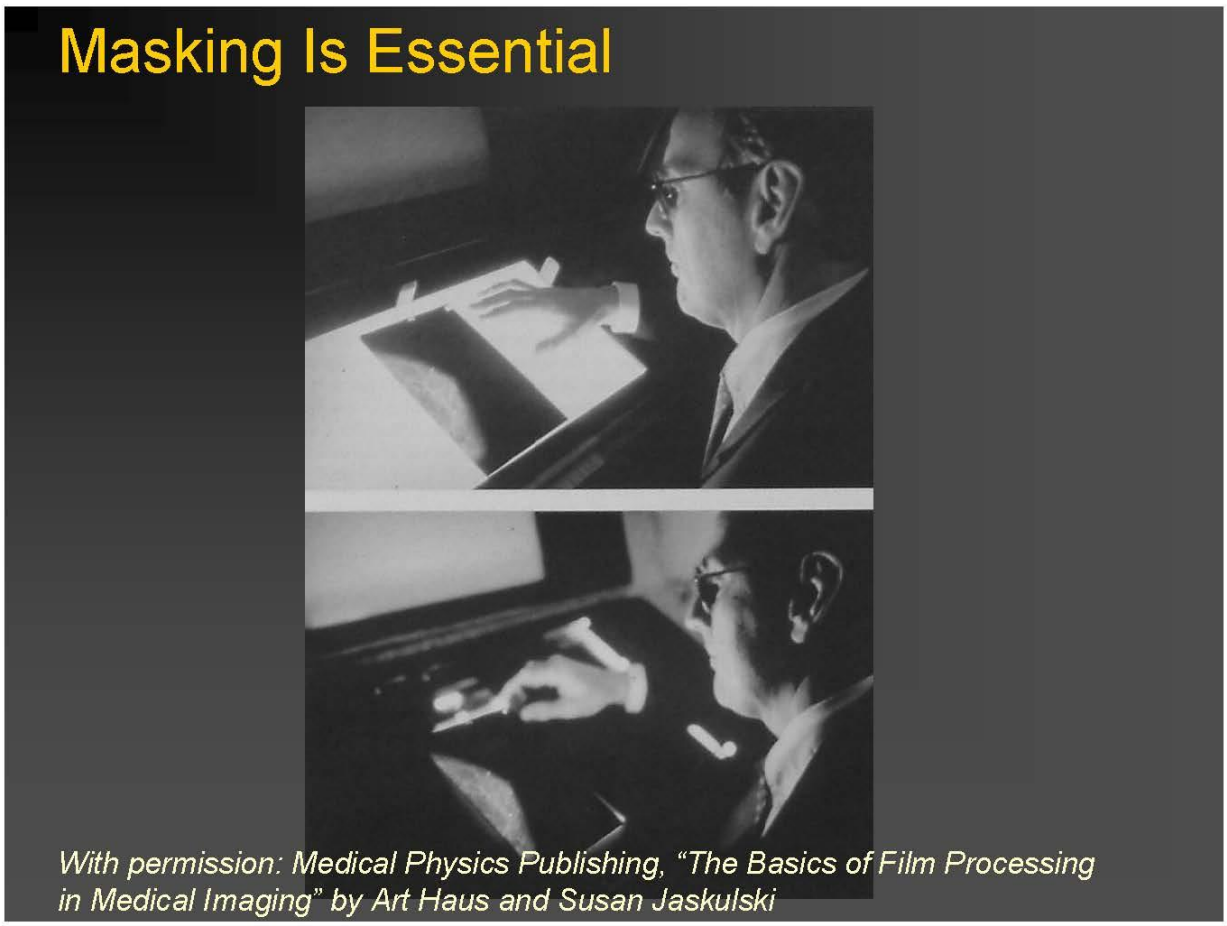


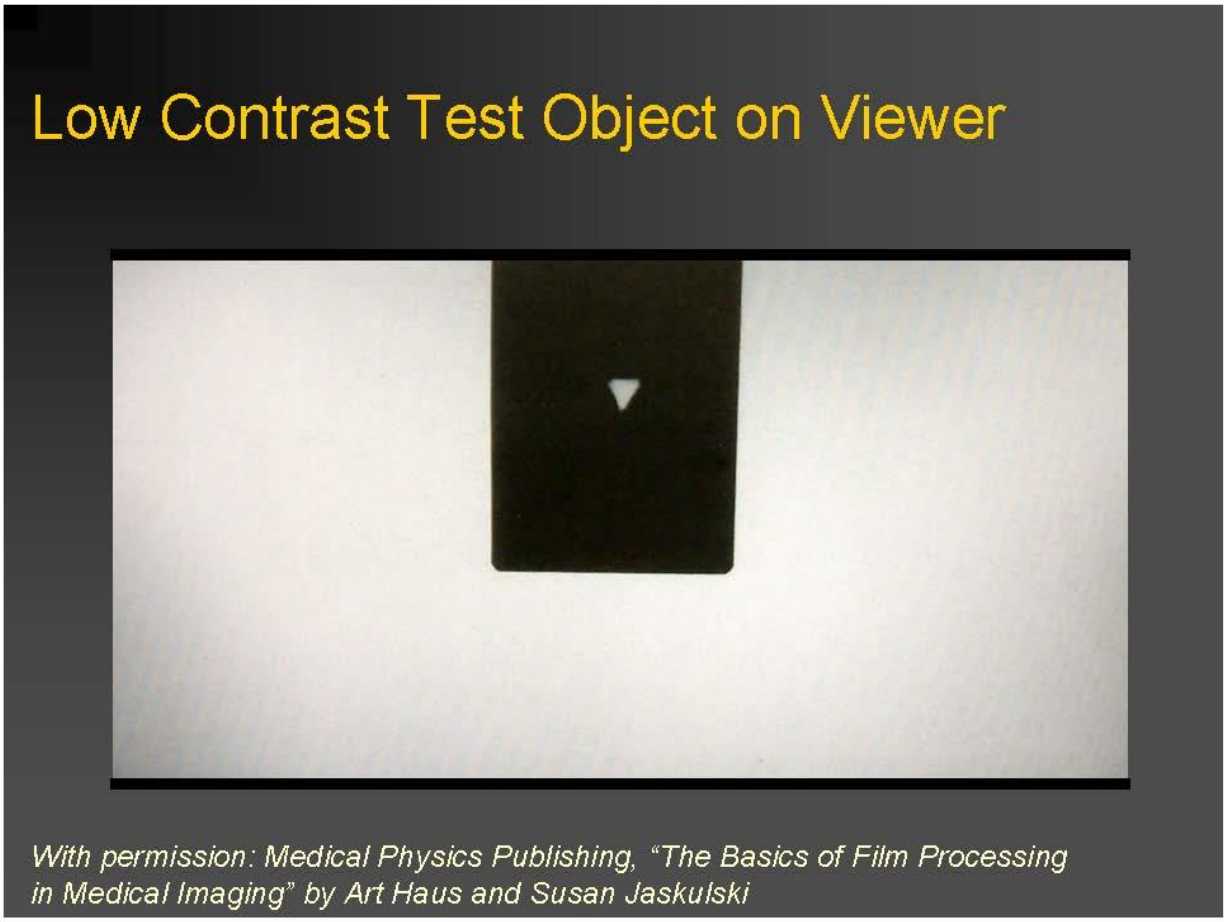
Processing Chemistry

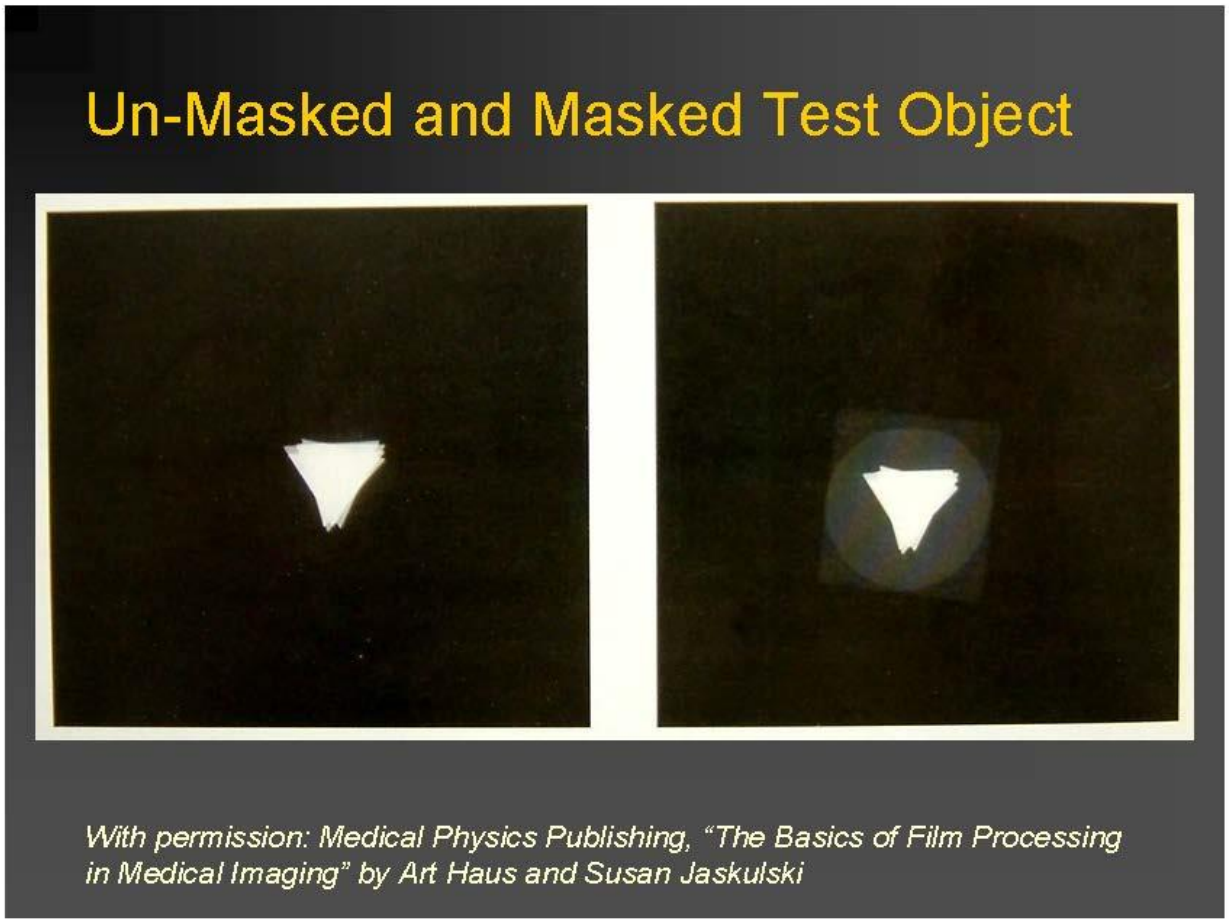
MQSA regulations require a facility to use chemical solutions that are capable of developing the films in a manner equivalent to film manufacturer's specifications.




ACR Film Viewing Recommendations
View box luminance ~ 3000 cd/m2
Masking is essential to preserve visibility of low contrast objects
Ambient light intensity < ~20 lux
High intensity spot light should be available
Magnifying glass should be available


Masking Is Essential
With permission: Medical Physics Publishing, "The Basics of Film Processing in Medical Imaging" by Art Haus and Susan Jaskulski


Low Contrast Test Object on Viewer
With permission: Medical Physics Publishing, "The Basics of Film Processing in Medical Imaging" by Art Haus and Susan Jaskulski


Un-Masked and Masked Test Object
With permission: Medical Physics Publishing, "The Basics of Film Processing in Medical Imaging" by Art Haus and Susan Jaskulski

D_{gN} Conversion (mrad / R)

Mo target / Mo filter

4.5 cm breast: 50% glandular and 50% adipose breast tissue composition

kVp

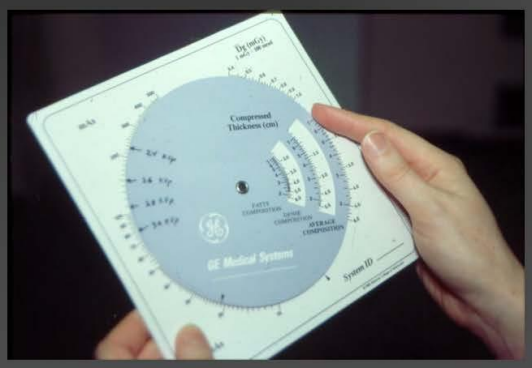
HVL (mm)	25	26	27	28	29	30	31	32
0.25	122							
0.26	126	128						
0.27	130	132	134					
0.28	134	136	138	139				
0.29	139	141	142	143	144			
0.30	143	145	146	147	148	149		
0.31	147	149	150	151	152	153	154	
0.32	151	153	154	155	156	158	159	160
0.33	155	157	158	159	160	162	163	164
0.34	160	161	162	163	164	166	167	168
0.35	164	166	167	168	169	170	171	172
0.36	168	170	171	172	173	174	175	176
0.37		174	175	176	177	178	178	179
0.38			179	180	181	182	182	183
0.39				184	185	186	186	187
0.40					189	190	191	192

ACR QC Manual 1999

Short-cut to Find MGD for MAP Phantom

MGD (mrad) = 0.5 x hvl (mm) x ESE (mR)

Short-cut gives MGD within 2-3% for all target and filters.



D. Jacobson, Radiographic exposure calculator and mammographic dose calculator, Radiology 1992; 182: 578.

INFORMATION FOR AUTHORS

PUBLICATION OF DOCTORAL THESIS AND DISSERTATION ABSTRACTS

A special feature of Medical Physics International (on line at www.mpijournal.org) is the publication of thesis and dissertation abstracts for recent graduates, specifically those receiving doctoral degrees in medical physics or closely related fields in 2010 or later. This is an opportunity for recent graduates to inform the global medical physics community about their research and special interests.

Abstracts should be submitted by the author along with a letter/message requesting and giving permission for publication, stating the field of study, the degree that was received, and the date of graduation. The abstracts must be in English and no longer than 2 pages (using the MPI manuscript template) and can include color images and illustrations. The abstract document should contain the thesis title, author's name, and the institution granting the degree.

Complete information on manuscript preparation is available in the INSTRUCTIONS FOR AUTHORS section of the online journal: www.mpijournal.org.

For publication in the next edition abstracts must be submitted not later than Април 1, 2019.

INSTRUCTIONS FOR AUTHORS

The goal of the new IOMP Journal Medical Physics International (http://mpijournal.org) is to publish manuscripts that will enhance medical physics education and professional development on a global basis. There is a special emphasis on general review articles, reports on specific educational methods, programs, and resources. In general, this will be limited to resources that are available at no cost to medical physicists and related professionals in all countries of the world. Information on commercial educational products and services can be published as paid advertisements. Research reports are not published unless the subject is educational methodology or activities relating to professional development. High-quality review articles that are comprehensive and describe significant developments in medical physics and related technology are encouraged. These will become part of a series providing a record of the history and heritage of the medical physics profession.

A special feature of the IOMP MPI Journal will be the publication of thesis and dissertation abstracts for will be the publication of thesis and dissertation abstracts for recent doctoral graduates, specifically those receiving their doctoral degrees in medical physics (or closely related fields) in 2010 or later.

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Manuscripts shall be in English and submitted in WORD. Either American or British spelling can be used but it must be the same throughout the manuscript. Authors for whom English is not their first language are encouraged to have their manuscripts edited and checked for appropriate grammar and spelling. Manuscripts can be up to 10 journal pages (approximately 8000 words reduced by the space occupied by tables and illustrations) and should include an unstructured abstract of no more than 100 words.

The style should follow the template that can be downloaded from the website at:

http://mpijournal.org/authors_submitapaper.aspx

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Illustrations can be inserted into the manuscript for the review process but must be submitted as individual files when a manuscript is accepted for publication.

The use of high-quality color visuals is encouraged. Any published visuals will be available to readers to use in their educational activities without additional approvals.

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Websites that relate to the manuscript topic and are sources for additional supporting information should be included and linked from within the article or as references.

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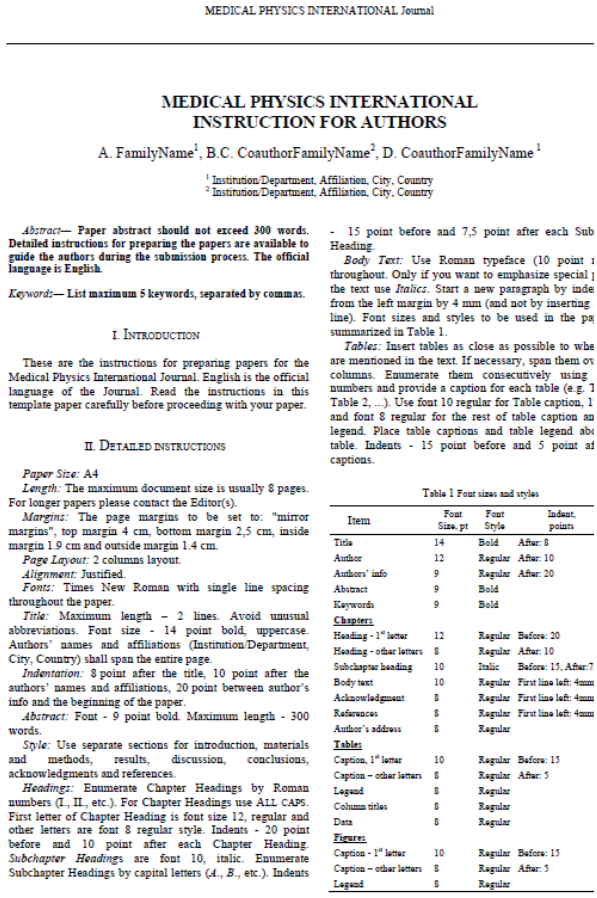
Authors considering the development of a manuscript for a Review Article can first submit a brief proposal to the editors. This should include the title, list of authors, an abstract, and other supporting information that is appropriate. After review of the proposal the editors will consider issuing an invitation for a manuscript. When the manuscript is received it will go through the usual peer-review process.



MEDICAL PHYSICS INTERNATIONAL Journal

MEDICAL PHYSICS INTERNATIONAL INSTRUCTION FOR AUTHORS

A. FamilyName[1], B.C. CoauthorFamilyName[2], D. CoauthorFamilyName [1]

[1] Institution/Department, Affiliation, City, Country
[2] Institution/Department, Affiliation, City, Country

Abstract— **Paper abstract should not exceed 300 words. Detailed instructions for preparing the papers are available to guide the authors during the submission process. The official language is English.**

Keywords— **List maximum 5 keywords, separated by commas.**

I. INTRODUCTION

These are the instructions for preparing papers for the Medical Physics International Journal. English is the official language of the Journal. Read the instructions in this template paper carefully before proceeding with your paper.

II. DETAILED INSTRUCTIONS

Paper Size: A4
Length: The maximum document size is usually 8 pages. For longer papers please contact the Editor(s).
Margins: The page margins to be set to: "mirror margins", top margin 4 cm, bottom margin 2,5 cm, inside margin 1.9 cm and outside margin 1.4 cm.
Page Layout: 2 columns layout.
Alignment: Justified.
Fonts: Times New Roman with single line spacing throughout the paper.
Title: Maximum length – 2 lines. Avoid unusual abbreviations. Font size - 14 point bold, uppercase. Authors' names and affiliations (Institution/Department, City, Country) shall span the entire page.
Indentation: 8 point after the title, 10 point after the authors' names and affiliations, 20 point between author's info and the beginning of the paper.
Abstract: Font - 9 point bold. Maximum length - 300 words.
Style: Use separate sections for introduction, materials and methods, results, discussion, conclusions, acknowledgments and references.
Headings: Enumerate Chapter Headings by Roman numbers (I., II., etc.). For Chapter Headings use ALL CAPS. First letter of Chapter Heading is font size 12, regular and other letters are font 8 regular style. Indents - 20 point before and 10 point after each Chapter Heading. *Subchapter Headings* are font 10, italic. Enumerate Subchapter Headings by capital letters (*A.*, *B.*, etc.). Indents - 15 point before and 7,5 point after each Sub Heading.
Body Text: Use Roman typeface (10 point throughout. Only if you want to emphasize special the text use *Italics*. Start a new paragraph by inde from the left margin by 4 mm (and not by inserting line). Font sizes and styles to be used in the pa summarized in Table 1.
Tables: Insert tables as close as possible to whe are mentioned in the text. If necessary, span them ov columns. Enumerate them consecutively using numbers and provide a caption for each table (e.g. T Table 2, ...). Use font 10 regular for Table caption, 1 and font 8 regular for the rest of table caption an legend. Place table captions and table legend abo table. Indents - 15 point before and 5 point af captions.

Table 1 Font sizes and styles

Item	Font Size, pt	Font Style	Indent, points
Title	14	Bold	After: 8
Author	12	Regular	After: 10
Authors' info	9	Regular	After: 20
Abstract	9	Bold	
Keywords	9	Bold	
Chapters			
Heading - 1st letter	12	Regular	Before: 20
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Subchapter heading	10	Italic	Before: 15, After:7
Body text	10	Regular	First line left: 4mm
Acknowledgment	8	Regular	First line left: 4mm
References	8	Regular	First line left: 4mm
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Tables			
Caption, 1st letter	10	Regular	Before: 15
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Column titles	8	Regular	
Data	8	Regular	
Figures			
Caption - 1st letter	10	Regular	Before: 15
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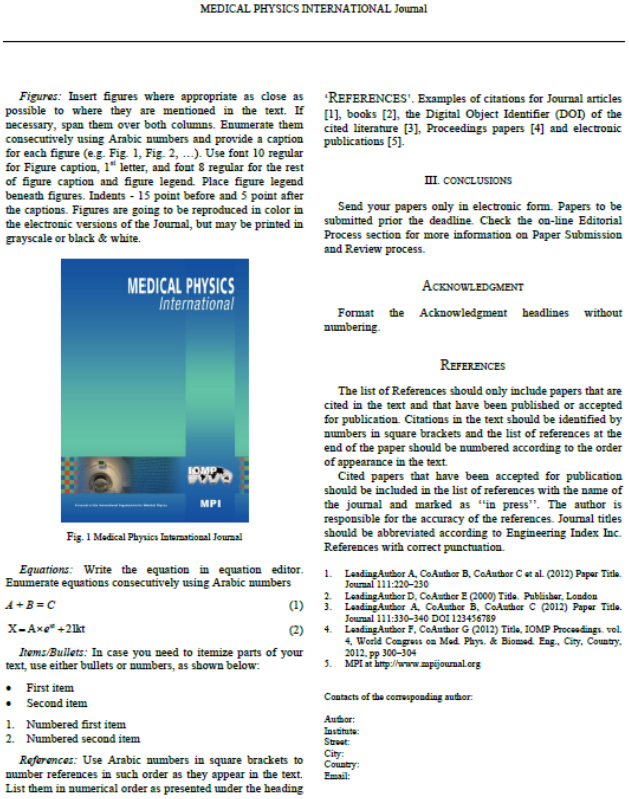
Legend	8	Regular	



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Figures: Insert figures where appropriate as close as possible to where they are mentioned in the text. If necessary, span them over both columns. Enumerate them consecutively using Arabic numbers and provide a caption for each figure (e.g. Fig. 1, Fig. 2, ...). Use font 10 regular for Figure caption, 1st letter, and font 8 regular for the rest of figure caption and figure legend. Place figure legend beneath figures. Indents - 15 point before and 5 point after the captions. Figures are going to be reproduced in color in the electronic versions of the Journal, but may be printed in grayscale or black & white.

Fig. 1 Medical Physics International Journal

Equations: Write the equation in equation editor. Enumerate equations consecutively using Arabic numbers

$$A + B = C \quad (1)$$

$$X = A \times e^{at} + 2lkt \quad (2)$$

Items/Bullets: In case you need to itemize parts of your text, use either bullets or numbers, as shown below:

- First item
- Second item

1. Numbered first item
2. Numbered second item

References: Use Arabic numbers in square brackets to number references in such order as they appear in the text. List them in numerical order as presented under the heading 'REFERENCES'. Examples of citations for Journal articles [1], books [2], the Digital Object Identifier (DOI) of the cited literature [3], Proceedings papers [4] and electronic publications [5].

III. CONCLUSIONS

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