MEDICAL RADIOLOGY

Diagnostic Imaging

Editors:
A. L. Baert, Leuven
M. F. Reiser, München
H. Hricak, New York
M. Knauth, Göttingen

U. Bick · F. Diekmann (Eds.)

Digital Mammography

With Contributions by

C. Balleyguier · E. Bellon · U. Bick · H. Bosmans · T. Deprez · F. Diekmann · C. Dromain
J. Jacobs · N. Karssemeijer · E. A. Krupinski · J. M. Lewin · G. Marchal · R. M. Nishikawa
P. Skaane · P. R. Snoeren · R. Van Engen · C. Van Ongeval · A. Van Steen · M. J. Yaffe
K. C. Young · F. Zanca

Foreword by

A. L. Baert

With 132 Figures in 224 Separate Illustrations, 32 in Color and 16 Tables

Springer

Ulrich Bick, MD
Charité
Campus Mitte
Institut für Radiologie
Charitéplatz 1
10117 Berlin
Germany

Felix Diekmann, MD
Charité
Klinik für Strahlenheilkunde
Augustenburger Platz 1
13353 Berlin
Germany

Medical Radiology · Diagnostic Imaging and Radiation Oncology
Series Editors:
A. L. Baert · L. W. Brady · H.-P. Heilmann · H. Hricak · M. Knauth · M. Molls · C. Nieder
M. F. Reiser

Continuation of Handbuch der medizinischen Radiologie
Encyclopedia of Medical Radiology

ISBN 978-3-540-78449-4 e-ISBN 978-3-540-78450-0

DOI: 10.1007/978-3-540-78450-0

Springer Heidelberg Dordrecht London New York

Medical Radiology · Diagnostic Imaging and Radiation Oncology ISSN 0942-5373

Library of Congress Control Number: 2009933608

Cover design: Publishing Services Teichmann, 69256 Mauer, Germany

Printed on acid-free paper

9 8 7 6 5 4 3 2 1

Springer is part of Springer Science+Business Media (www.springer.com)

Foreword

Digital Radiography has been firmly established in diagnostic radiology during the last decade. Because of the special requirements of high contrast and spatial resolution needed for roentgen mammography, it took some more time to develop digital mammography as a routine radiological tool.

Recent technological progress in detector and screen design as well as increased experience with computer applications for image processing have now enabled *Digital Mammography* to become a mature modality that opens new perspectives for the diagnosis of breast diseases.

The editors of this timely new volume Prof. Dr. U. Bick and Dr. F. Diekmann, both well-known international leaders in breast imaging, have for many years been very active in the frontiers of theoretical and translational clinical research, needed to bring digital mammography finally into the sphere of daily clinical radiology.

I am very much indebted to the editors as well as to the other internationally recognized experts in the field for their outstanding state of the art contributions to this volume. It is indeed an excellent handbook that covers in depth all aspects of *Digital Mammography* and thus further enriches our book series *Medical Radiology*. The highly informative text as well as the numerous well-chosen superb illustrations will enable certified radiologists as well as radiologists in training to deepen their knowledge in modern breast imaging. Also gynecologists, oncologists, and breast surgeons will find it a very valuable instrument for better therapeutic management of their patients with breast disease.

I congratulate the editors with their fine contribution to the medical literature and I am convinced that their book will meet great success with our readership.

Albert L. Baert
Series Editor

Preface

The first article on X-ray examinations and breast cancer was published by A. Salomon, a surgeon at Charité university hospital in Berlin, nearly a hundred years ago. Although mammography has come a long way since then, there is an ongoing debate about its limitations. A major issue is that mammography alone does not allow reliable detection of breast cancer, especially in women with dense glandular tissue.

Full-field digital mammography – first introduced almost exactly 10 years ago – not only significantly facilitates different aspects of the workflow in mammography such as image viewing, communication, and storage, but also promises to improve image quality by enhancing contrast in dense areas of the breast.

A thorough understanding of the capabilities of this new technology is important for its users and their clinical partners alike. This book discusses the physical and technical concept of digital mammography along with clinical aspects such as differences between film-screen and digital mammography in visibility and appearance of mammographic findings. Digital mammography is a rapidly evolving imaging modality including novel and emerging mammographic techniques such as contrast-enhanced digital mammography and digital breast tomosynthesis.

Many renowned international experts in the field of digital mammography – both physicists and physicians – have contributed to this book. We believe that this mixture will provide a deeper insight into digital mammography and its current role.

We thank all authors and the staff of Springer for their patient and tireless support during the preparation of this book.

Berlin, Germany

Ulrich Bick
Felix Diekmann

Contents

Abbreviations

AAPM	American Association of Physicists In Medicine
ACR	American College of Radiology
ACRIN	American College of Radiology Investigational Network
AEC	Automatic exposure control
AGD	Average glandular dose
ANOVA	Analysis of variance
AUC	Area under the receiver operating characteristic curve
BIRADS	Breast Imaging Reporting and Data System
CAD	Computer-aided diagnosis; refers to the field as a whole and includes both CADe and CADx
CADe	Computer-aided detection; detection of lesions; used in screening mammography
CADx	Computer-aided diagnosis; classifying lesions as benign or malignant; used in diagnostic mammography
CC	Cranio-caudal
CCD	Charge-coupled device
CE-DBT	Contrast-enhanced digital breast tomosynthesis
CEDM	Contrast-enhanced digital mammography
CELBSS	Central and East London Breast Screening Service
CLAHE	Contrast limited adaptive histogram equalization
CNR	Contrast-to-noise ratio
CR	Computed radiography
d	Spot aperture
DBT	Digital breast tomosynthesis
DCIS	Ductal carcinoma in situ
Del	Detector element
DICOM	Digital imaging and communications in medicine
DMIST	Digital mammography imaging screening trial
DQE	Detective quantum efficiency
DR	Direct Radiography Systems
DSPP	Digital screening project preventicon
E	Energy
EC	European Commission
EMR	Electronic medical record
Eqn	Equation
EUREF	European Reference Organisation for Quality Assured Breast Screening
FDA	United States Food and Drug Administration
FFDM	Full-field digital mammography
FN	False negative
FPF	False-positive fraction
FROC	Free-response receiver operating characteristic
GSDF	Grayscale standard display function
η	Quantum detection efficiency
H&D	Hurter-Driffield curve
HIP	Health Insurance Program of New York Project
HIS	Hospital information system
HIW	Histogram-based intensity windowing

IAEA	International Atomic Energy Authority
IDC	Invasive ductal carcinoma
IHE	Integrated Healthcare Enterprise
IHE SUP	IHE mammo specific supplement
IHE TF	IHE technical framework
IHE WB	IHE mammo workbook
INBSP	Irish National Breast Screening Program
IP	Imaging plate
IPEM	Institute of Physics and Engineering in Medicine
JAFROC	Jack-knife free-response receiver operating characteristic
λ	Wavelength
LROC	Localization response operating characteristic
LUT	Lookup tables
MDM	Micro dose mammography
MIP	Maximum intensity projection
ML	Medio-lateral
ML-EM	Maximum likelihood expectation maximization
MLO	Medio-lateral oblique
MoniQA	Certain test pattern for quality control
MPPS	Modality performed procedure step
MQSA	Mammography Quality Standards Act
MTF	Modulation transfer function
Mx	Mammography
n(E)	Quantum interaction efficiency
n_a	X-rays interacting with the detector
NBCSP	Norwegian Breast Cancer Screening Program
n_d	Certain mean number of x rays
NEQ	Noise equivalent quanta
NHS	National Health System
NHSBSP	National Health Service Breast Screening Program
NLF	Non-lesion localization fraction
NPS	Noise power spectra
NPV	Negative predictive value
NS	Required number of signal steps
OD	Optical density
P	Centre-to-centre distance or pitch
PACS	Picture archiving and information systems
pAUC	Partial area under the receiver operating characteristic curve
PERFORMS 2	Personal Performance in Mammography Screening
PMMA	Polymethylmethacrylate
PPV	Positive-predictive value
PPV	Positive predictive value
PSP	Photostimulable Phosphor
QA	Quality Assurance
QC	Quality Control
RDOG	Radiology Diagnostic Oncology Group
RIS	Radiology Information System
ROC	Receiver operating characteristic
ROI	Region of interest
RTOG	Radiation Therapy Oncology Group
SART	Simultaneous algebraic reconstruction technique
SD	Standard deviation
SDNR	Signal difference-to-noise ratio
SFM	Screen-film mammography
SMF	Standard mammogram form
SNR	Signal-to-noise ratio

SWOG	Southwest Oncology Group
TDI	Time-delay integration
TFT	Thin film transistor
TG18	AAPM Topic Group 18
TN	True negative
TPF	True-positive fraction
VOI LUT	Values of interest lookup table
VPN	Virtual private network
w	Inverse amount of signal produced upon interaction for a particular X-ray energy
n_B	Number of x rays that are transmitted along path B
C_{rad}	Radiation contrast
σ^2_{tot}	Overall image noise
n_0	Absorbed radiation dose
n_{sq}	Number of secondary light quanta or electrons

Basic Physics of Digital Mammography

1

Martin J. Yaffe

CONTENTS

Martin J. Yaffe, PhD
Imaging Research Program, Sunnybrook Health Sciences Centre, University of Toronto, 2075 Bayview Avenue, Toronto, ON, Canada M4N 3M5

KEY POINTS

Digital mammography overcomes several technical limitations associated with screen-film mammography. An essential feature of digital mammography is that both the intensity and the spatial distribution of the X-ray transmission pattern are sampled to form the image. In the spatial domain, the interval between samples (pitch) and the response profile of the detector element (del) largely determine the spatial resolution of the imaging system. The dynamic range of the detector and the number of bits used to digitize the image determine the ability to image all parts of the breast with acceptable contrast and signal-to-noise ratio. Depending on the system design, it is possible to eliminate much of the structural or fixed-pattern noise associated with the detector and the X-ray beam to approach a quantum noise limited situation. In digital mammography systems, it is often possible to design detectors that allow efficient use of the incident X-rays without excessive loss of spatial resolution. This permits a substantial reduction in the radiation dose to the breast when compared with film mammography without sacrifice of image quality. Because of the differences in technology, the optimum exposure conditions may shift toward the use of higher energy spectra than would be used with film, particularly for dense or thick breasts.

1.1 Introduction

There are several key features of digital mammography that distinguish it from screen-film mammography and contribute to its potential advantages. Probably, the most significant property of digital mammography is that it decouples the processes of image acquisition from the subsequent stages of archiving, retrieval, and image display. Unlike the situation in film mammography where these processes are inextricably linked, this facilitates optimization of each of the separate functions and great flexibility in the adjustment of image display characteristics. Because the image data are captured in numerical form, this implies that the image data are sampled both spatially and in the signal level. Sampling must be carried out appropriately because undersampling can cause potential limitations to image quality while oversampling can reduce the efficiency and increase the cost of operating the digital mammography system.

A well-optimized digital mammography system can provide the following benefits:

1. More efficient acquisition of the X-ray data for the mammogram because
   (a) The detector can be made thick enough to absorb a large fraction of the X-rays transmitted by the breast
   (b) Elimination of granularity noise
   (c) Reduction of radiation dose
2. Capture of the image data in numerical form
3. Control of display brightness and contrast that is independent from the amount or characteristics of the X-ray exposure
4. Image processing to adapt the image to match visual performance of the eye and overcome limitations of the display device
5. Ability to remove other structural (fixed-noise) patterns by flat-field correction
6. Quantitative imaging techniques, telemammography, CAD, tomosynthesis, contrast imaging

1.2 Characterizing Imaging Performance

To evaluate imaging systems or to compare the performance of a novel system to a conventional imaging device, it is necessary to have quantitative performance measures. Important imaging parameters to be considered are contrast, spatial resolution, noise characteristics, and dynamic range. While a detailed treatment of these is beyond the scope of this book and is available elsewhere (e.g., Barrett and Myers 2001), the essential descriptors of imaging performance such as radiation contrast, signal-to-noise ratio, MTF, DQE, and noise-equivalent quanta will be briefly introduced in the process of discussing the basic physics of digital mammography.

1.3 Basic Physics of Image Acquisition

It is instructive to analyze the key elements of the physics of X-ray image acquisition in mammography through a simple model of a breast containing a structure of interest (Fig. 1.1). This structure could be a tumor, a microcalcification, or some normal aspect of the breast anatomy.

For a monoenergetic X-ray beam, the mean number of X-rays transmitted along Path A through normal breast tissue and arriving at a hypothetical plane beyond the breast, referred to as the image plane, is:

$$n_A = n_0 e^{-\mu z} \tag{1.1}$$

where n_0 is the mean number of X-rays incident on the breast, z is its thickness, and μ is the X-ray attenuation coefficient of the tissue. Here, the divergence of X-rays from a point source has been ignored and the simplifying assumption has been made that no scattered radiation reaches the image plane. The number of X-rays that are transmitted along Path B passing

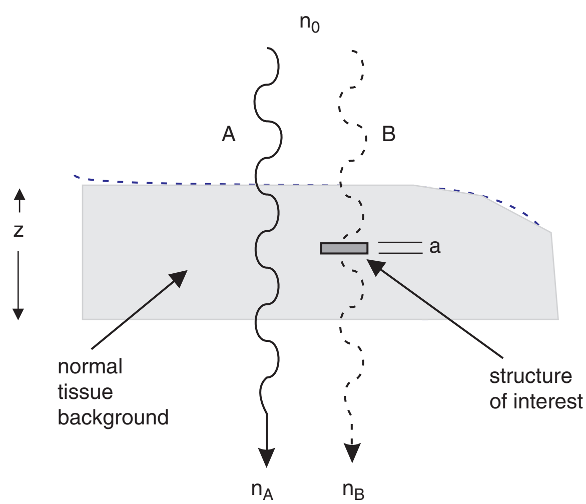

Fig. 1.1. Schematic diagram of the breast illustrating the basic imaging problem of detecting differences in X-ray transmission between Path A passing through normal tissue and Path B passing through a region containing a structure of interest such as a lesion in a breast of varying thickness (from Pisano et al. 2004. With permission)

through the structure of interest in the breast having X-ray linear attenuation coefficient, μ' is:

$$n_B = n_0 e^{-\mu(z-a)-\mu' a} \tag{1.2}$$

where a is the thickness of the structure in the direction of travel of the X-rays. The signal difference produced by the presence of the structure is

$$\mathrm{SD} = n_A - n_B \tag{1.3}$$

The resultant radiation contrast can be defined as:

$$C_{\mathrm{rad}} = \frac{n_A - n_B}{n_A + n_B} \tag{1.4}$$

Substituting (1.1) and (1.2) into (1.4), one obtains:

$$C_{\mathrm{rad}} = \frac{1 - e^{-(\mu-\mu')a}}{1 + \varepsilon^{-(\mu-\mu')\alpha}} \tag{1.5}$$

This expression demonstrates that the radiation contrast is determined by two factors, the difference in attenuation coefficient between the background breast tissue and the structure and the thickness of the structure. Note that in this simplified model, the contrast does not depend on n_0, z, or μ. In practice, where the X-ray spectrum is polyenergetic, and where some scattered radiation is recorded, C_{rad} will show some dependence on these variables.

The breast is composed primarily of fat and fibroglandular tissue. Measured attenuation coefficients of these materials as well as those of breast tumor specimens are plotted in Fig. 1.2. Note that both the attenuation coefficients of these materials and the difference in μ between any pair of materials decreases with increasing X-ray energy. Therefore, radiation contrast will decrease with increasing X-ray energy. The need to achieve an adequate radiation contrast is the reason why relatively low X-ray energies are employed in mammography. In Fig. 1.3, radiation contrast is plotted vs. energy for a modeled breast composed of 30% fibroglandular tissue and 70% fat and containing a 3-mm-thick tumor.

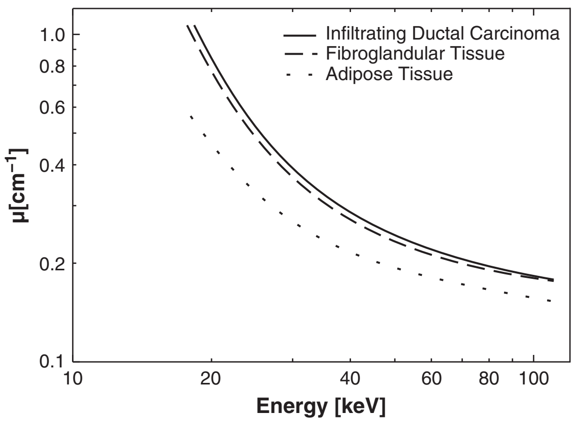

Fig. 1.2. Measured linear X-ray attenuation coefficients of fat, fibroglandular tissue, and tumor in the breast (from Johns and Yaffe (1987). With permission, IOP publications)

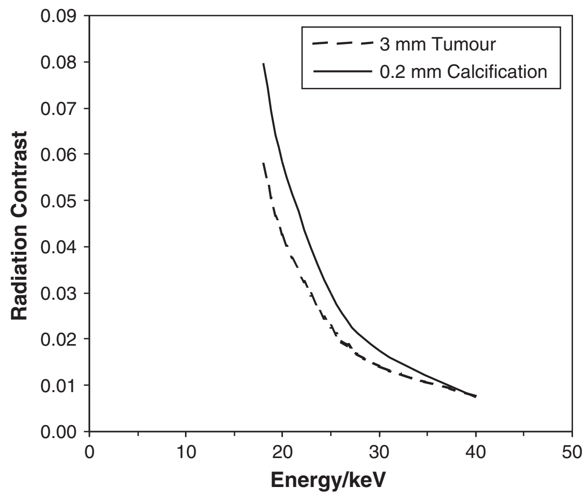

Fig. 1.3. Dependence on contrast of a breast mass and a calcification on X-ray energy. In this example, the breast is composed of 70% fat and 30% fibroglandular tissue. The tumor is modeled as being 3 mm thick, and the calcification is 0.2 mm thick

1.3.1 Detection of X-Rays

To form an image, the transmitted X-rays must be detected and their energy converted into a usable signal. Quantum detection efficiency, η, describes the fraction of the X-rays incident on the detector, that interact with it, producing at least some signal. The quantum detection efficiency is given by:

$$\eta(E) = 1 - e^{-\mu(E)d} \tag{1.6}$$

where $\mu(E)$ is the X-ray linear attenuation coefficient of the detector material, which depends on the X-ray energy, E, and d is the thickness of the active region of the detector, i.e., the region from which the signal is produced. The quantum detection efficiency increases with increasing d and μ. The value of μ depends on the density and atomic number of the absorber.

Properties of X-ray detectors for digital mammography will be discussed further in Chap. 2.

With reference to Fig. 1.1, the actual number of X-rays that will be detected for Paths A and B is $n_{dA} = \eta n_A$ and $n_{dB} = \eta n_B$, respectively.

1.3.2 Recording of the Image

There are fundamental differences between digital and film mammography in the way that the information carried by the detected X-rays is transferred and managed to form an image. In a film system, the intensifying screen produces an amount of light that is directly proportional to the amount of energy deposited by the X-rays. This exposes the film, which is subsequently chemically processed to produce a pattern of optical density, which comprises the image. The transfer characteristics (often referred to as the Hurter–Driffield curve) of a mammography screen–film combination are shown in Fig. 1.4a. In this plot of optical density (blackness of the processed film) versus the logarithm of relative X-ray exposure to the screen, it is seen that the response is highly nonlinear and it tends to flatten for exposures above and below a fairly restricted range. This limited range has important implications on image quality. The gradient or slope of this curve defines the amount by which the radiation contrast, C_{rad}, is either amplified or diminished in displaying the image. The value of optical density determines the brightness with which the image will appear on the viewbox. Where the curve becomes flat, the displayed contrast will be poor. The film characteristic imposes a compromise between the displayed contrast and the latitude or range of radiation over which the contrast is acceptable. Once a film image has been exposed, the display contrast characteristics are fixed and, if the image is of inadequate brightness or contrast, it must be re-exposed. For this reason, automatic exposure control (AEC) in mammography is very critical. The AEC attempts to terminate a film exposure at a point when the tissue above the sensor in the AEC has transmitted an appropriate number of X-rays to expose the film to a level where the gradient will be at or near its maximum value and the viewing brightness of the image is acceptable. Of necessity, the image corresponding to other areas of the breast may be sub-optimally exposed.

The characteristic curve of a typical digital mammography detector is shown in Fig. 1.4b. The detector inherently produces a signal that is linearly proportional to the intensity of X-rays transmitted by the breast. It has a very large dynamic range,[1] so that it is possible to produce a faithful representation of X-ray transmission for all parts of the breast. Furthermore, unlike film, the shape of the characteristic curve for these detectors is much less dependent on the level of radiation exposing the detector. Maintaining an adequate value of C_{rad} is still important; however, both the displayed image brightness and contrast as well as other viewing characteristics can be adjusted separately at a computer console during image viewing. Image display will be discussed in detail in Chap. 7.

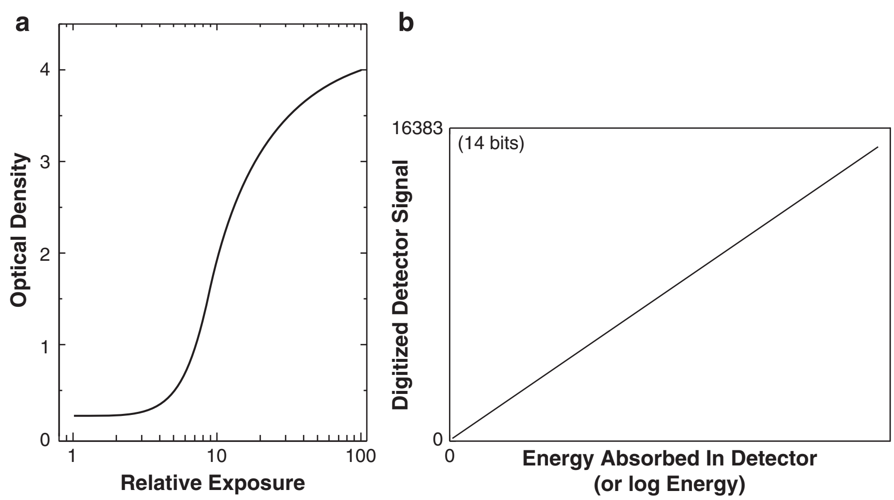

Fig. 1.4. (**a**) Characteristic (H and D) curve of a mammographic screen-film system. Optical density (OD) of the processed film is plotted versus the log of the relative X-ray exposure to the fluorescent intensifying screen. (**b**) Characteristic response of a detector designed for digital mammography (from PISANO et al. 2004. With permission)

[1]Dynamic range is loosely defined as the range of exposures over which the detector provides a reliably measurable signal. Some of the factors affecting dynamic range of the imaging system are discussed in the next section on sampling.

An AEC is important in determining the exposure level to the detector and to the breast, but its key roles are to assist in achieving a predetermined signal-to-noise ratio (see next section) and a reasonable radiation dose to the breast rather than determining the image brightness or contrast.

1.3.3 Sampling

A basic difference between the detectors used for screen-film mammography and for digital mammography is that in the former, the signal varies more or less continuously in both the spatial and intensity domains as in Fig. 1.5a, while in digital systems, the analog signal from the detector is sampled. Sampling occurs both spatially and in quantization of signal levels. Spatially, the digital image is represented by a matrix whose elements (pixels) are considered to be independent from one another. In an image, an analog-to-digital converter (Fig. 1.5b) is generally used to sample the analog signal from the detector such that it will be represented in each image pixel as an integer digital value (Fig. 1.5c). Owing to the sampling process, no meaningful spatial information in the image can be represented at a scale smaller than a pixel and no signal level can be subdivided more finely than the integers comprising the image data. Therefore, the pixel size and the interval between signal digitization levels can have an important influence on image spatial resolution and contrast sensitivity, respectively.

1.3.3.1 Some Spatial Sampling Concepts

As will be discussed in Chap. 2, different detectors use different approaches to achieve spatial sampling. Nevertheless, there are some important concepts that are common to all systems. Figure 1.6 illustrates part of a detector, divided into detector elements or dels. In image acquisition, each del provides one or (in the case of scanning systems) a series of discrete X-ray measurements to contribute to the image.

In the simplest case, the signal from one del will supply the information displayed in one pixel of the final image. The dels are arrayed with a given center-to-center distance or pitch, p. In addition, it is possible that only part of the del is actively sensitive to incoming X-rays, for example a square region of dimension d.

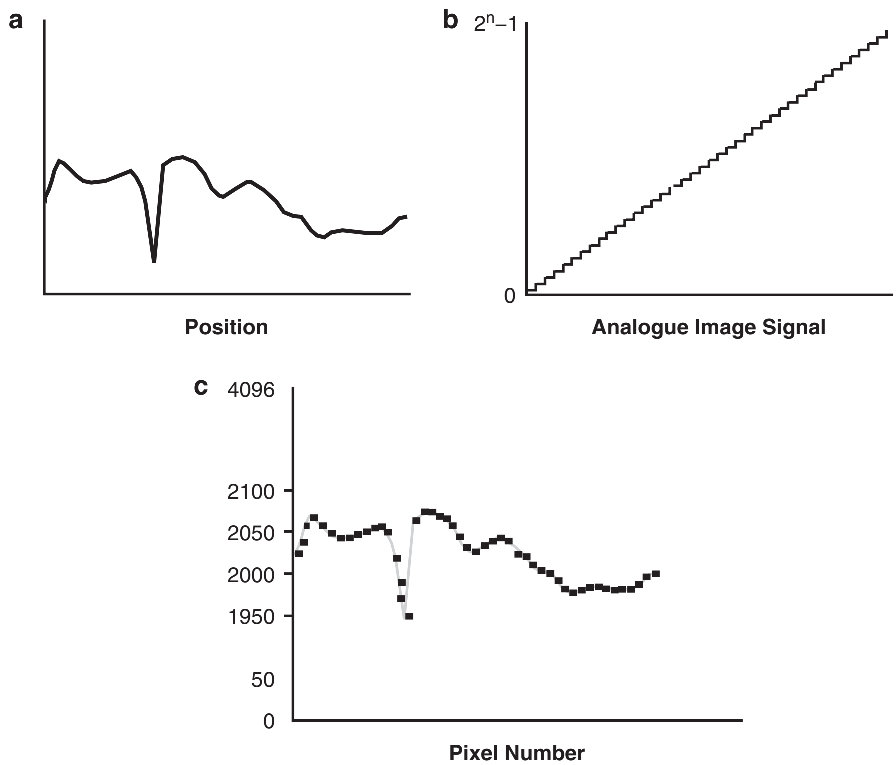

Fig. 1.5. Sampling of an X-ray pattern in the spatial and intensity domains to create a digital image. Here, only one of the two image dimensions is shown. Unlike the analog image (**a**), which is defined continuously in space and signal level, the digital image (**b**) is pixelated at discrete points and only a finite number of signal levels are recorded (from Pisano et al. 2004. With permission)

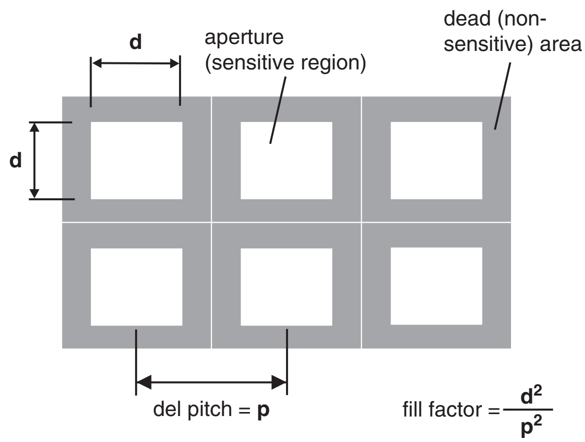

Fig. 1.6. Concept of the del and spatial sampling. A detector element (del) contains an active region with dimension d. Dels are spaced at a pitch p. Because of inactive detector material on the del, the fraction of the area that is sensitive to X-rays, d^2/p^2, also known as the "fill factor," can be less than 1 (reprinted from Haus and Yaffe et al. (2000). With permission from Elsevier)

Here, d is referred to as the aperture size. If d is smaller than p, the loss of X-ray signal due to this geometric waste will cause the efficiency of the detector and its sensitivity to be reduced to d^2/p^2 of the possible value for d = p. Here, d^2/p^2 is known as the fill factor. In addition to the effect on efficiency and sensitivity, the values of d and p affect other aspects of imaging performance and these will be discussed later.

1.3.3.2 Sampling of Signal Level

Both the detector and the digitization of signal must be designed to cover the appropriate range of X-ray fluence to accommodate the most radiolucent areas (near the periphery of the breast) and the most radio-opaque areas of any breast that is to be imaged. In addition, to provide excellent contrast sensitivity (the ability to distinguish between structures providing only subtle changes in X-ray attenuation), the smallest digitized signal of interest that will occur in a location corresponding to the thickest, densest part of the breast, must be measured to the required precision. Therefore, the required dynamic range of analog signal from the detector must be adequate. Ideally, the characteristics of this response will be constant over that range, i.e., a detector designed to have linear response with fluence will maintain that response over the entire range and likewise with a system that provides logarithmic response.

The digitization process represents the X-ray fluence by a binary number between 0 and 2^n-1, where n is referred to as the *number of bits* of digitization. Including 0, this provides NS = 2^n signal steps. When the signal is subsequently digitized, the number of bits must be adequate to cover both the range of attenuation of X-rays by the breast, R, and also provide the needed precision at the bottom end of the range. For a linear system, the required number of signal steps is then:

$$\mathrm{NS} = 100\,R/q \tag{1.7}$$

where q is the required precision in percent. For example, if the maximum attenuation imposed by the breast is a factor of 60 and the required precision is 1%, then NS = 6,000. Digitization would require at least 13 bits as 12 bits provides only 4,096 steps. In practice, a 14-bit digitizer has been found to be more than adequate for digital mammography.

1.4 Noise

1.4.1 Quantum Noise

Both the production of X-rays and their interaction in a detector occur in a random manner whose statistics can be described by the Poisson distribution. That means that even for a part of the breast whose X-ray attenuation was absolutely constant, if the average number of X-rays recorded over a particular image area is $\langle n \rangle$ this number will fluctuate from location to location with a standard deviation of $\sigma = \sqrt{\langle n \rangle}$. This fluctuation occurs in the absence of variations in attenuation of the breast and is referred to as *quantum noise*. It is important to realize that the value of σ is determined, not by the number of X-rays incident on the detector, but by how many are used to form the image, i.e., by the number, n_d, interacting with the detector. The *signal-to-noise ratio (SNR)* can be defined as the ratio of $\langle n_\mathrm{d} \rangle$ to σ and so is given by:

$$\mathrm{SNR} = \frac{\langle n_\mathrm{d} \rangle}{\sqrt{\langle n_\mathrm{d} \rangle}} = \sqrt{\langle n_\mathrm{d} \rangle} \tag{1.8}$$

Therefore, if it is desired to reduce the apparent noisiness of the image (i.e., increase the SNR) to allow the perception of more subtle features, the radiation level absorbed by the detector, n_d, must be increased. This could be accomplished in two ways: by increasing the exposure factors (i.e., mAs) or by employing a detector with an increased value of η.

1.4.2 Structural Noise

In film-based mammography systems, another important noise source is the random fluctuation contributed by the granularity of the film itself. The film emulsion is composed of grains of silver halide and their random structure increases the total noise. The overall image noise can be estimated by treating noise sources as being statistically independent, giving:

$$\sigma^2_{\text{tot}} = \sigma^2_{\text{q}} + \sigma^2_{\text{gr}} + \cdots \tag{1.9}$$

where σ_q is the quantum noise and σ_{gr} the film grain noise. The presence of sources of noise other than quantum fluctuation will reduce the SNR and it is therefore desirable to minimize them. Other noise sources associated specifically with detectors will be discussed in Chap. 2.

In film imaging, each sheet of film has a different pattern of structural granularity and, therefore, there is no practical way to remove the effect of this noise from the mammogram. In most digital mammography systems, film granularity is eliminated; however, there is usually some structural variation across the image field, much of which is associated with spatial variations in detector sensitivity. Because these generally remain constant over time, they do not really represent noise in the traditional sense of being truly random in space and time, but are instead referred to as "fixed pattern noise" or "structural noise." The availability of data from the detector in digital form allows the effects of structural noise to be removed by image correction. An approach to such a correction will be described in Chap. 2.

1.4.3 Signal Difference-to-Noise Ratio

For characterizing the quality of the image in terms of the potential detectability of structures in the breast, it is useful to introduce the signal difference-to-noise ratio (SDNR). The SDNR is simply the ratio of the signal difference, SD, from (1.3) (but using the detected rather than incident number of X-rays) to the standard deviation, σ_{tot},

$$\text{SDNR} = \frac{(n_{\text{Ad}} - n_{\text{Bd}})}{\sigma_{tot}} \tag{1.10}$$

The SDNR is related to the SNR and the radiation contrast by:

$$\text{SDNR} = 2C_{\text{rad}}\text{SNR} \tag{1.11}$$

Although both SNR and SDNR can be evaluated for single pixels in an image, it is generally more meaningful to consider these quantities over larger areas corresponding to a structure of interest or lesion and (in the case of SDNR) an equal area of adjacent background.

1.5 Radiation Dose

The absorbed radiation dose in the breast is proportional to, n_0, the number of X-ray quanta incident on the breast. If the required SNR is specified, this defines the number, n_d, of X-rays that must be detected (1.8), giving:

$$n_0 = \frac{n_d e^{\mu(E)z}}{\eta(E)} \tag{1.12}$$

The dose can then be determined by multiplying n_0 by the appropriate conversion factors (ICRU 2005).

1.6 Scattered Radiation

In mammography, some X-rays will pass through the breast without interaction, some will be absorbed, and some will scatter in the breast and escape. At mammographic energies, for an average breast, 40% or more of the X-rays directed toward the detector may have scattered one or more times in the breast (Barnes and Brezovich 1977, 1978). In projection radiography, scattered radiation is not considered to carry any useful information. There are several effects of scattered radiation on radiographic imaging; however, there are differences in the significance of these effects in digital and film mammography. Recording of scattered X-rays uses part of the dynamic range of the detector. In screen-film mammography where the dynamic range is already quite limited, this can be an important negative factor. In digital mammography, the detector generally has a sufficiently large dynamic range that this is not a concern.

Second, there is a fairly uniform haze that is imposed over the entire image. This reduces C_{rad}. Again, in film mammography, this effect can be significant while in digital mammography it is less so because image processing can be used to recover displayed contrast.

Finally, recording the scatter adds statistical fluctuation without information, thereby reducing the SNR. This is probably the major detrimental effect of scatter in digital mammography.

In screen-film mammography and in many digital mammography systems, scatter directed toward the image receptor is partially removed by an antiscatter grid. The grid is not efficient, in that it removes part (25–30%) of the useful directly transmitted "primary" X-ray beam, while rejecting most, but not all (80–90%), of the scattered radiation (Wagner 1991; De Almeida et al. 1999). The loss of both primary and scattered radiation reduces the number of X-rays recorded by the receptor. In film, both these must be replaced by increased exposure from the X-ray tube to ensure that the film is exposed to the proper level on its characteristic curve. The resultant increase in the tube output (and the radiation dose to the breast) is called the *Bucky factor* and this can be on the order of 2–2.5. In digital mammography, it is not necessary to compensate for the loss of scatter removed by the grid, as image brightness and contrast can be adjusted during image viewing. On the other hand, if there is to be no loss of SNR, then the exposure should be increased enough to make up for the loss of primary in the grid and maintain a desired value of n_d. In digital systems, it is also feasible to employ an alternative scanning approach to discriminate against scatter by acquiring the imaging in a geometry where the X-ray beam and detector are scanned across the breast and the beam is restricted to a slit or slot format. This is discussed further in Chap. 2.

1.7 Spatial Resolution

Before discussing the factors that affect spatial resolution in digital mammography, a brief introduction to the modulation transfer function, a valuable tool for quantifying resolution, is provided.

1.7.1 Modulation Transfer Function

In film-based imaging, spatial resolution is often assessed by determining the limiting resolution in terms of line-pairs/mm from a bar pattern. This is a subjective test, however, and is not very useful in the analysis of complex imaging systems.

Spatial resolution can be characterized quantitatively and more usefully through the *modulation transfer function* (MTF). The MTF describes how well the imaging system or one of its components such as the detector transfers the contrast of sinusoidal patterns from the incident X-ray pattern to the output. A sinusoid is a repetitive function, characterized as having a frequency (in this case a *spatial frequency* specified in cycles/mm) and an amplitude. The concept of spatial frequency can be visualized by considering ripples in a pond. Low spatial frequencies (long distance between wave peaks) represent coarse structures and high spatial frequencies (short wavelengths) describe fine detail.

Any pattern can be represented as a combination of sinusoidal shapes, each spatial frequency having a specific amplitude. The MTF simply describes how well each spatial frequency is transferred through a system, i.e., is simply the ratio of the amplitude of the sinusoid at the output of the system or a component to that at the input. The MTF of an imaging system is often 1.0 at very low spatial frequencies and falls with increasing spatial frequency. In a system containing several elements that affect the spatial resolution, the overall MTF is determined as the product of the MTFs of the individual components. For example, the MTF of a radiographic system is the product of that due to the focal spot, the detector, and any motion of the patient during the exposure. This is helpful in determining what part of the system is responsible for limiting its performance.

The MTF of a typical screen-film detector is shown in Fig. 1.7. As seen from the figure, it extends well beyond 20 cycles/mm. It is mainly determined by the screen, as the film has a very high MTF.

Several factors affect the spatial resolution in digital mammography. Some of them are identical to those that apply to film mammography, i.e., the focal spot must be sufficiently small to prevent excessive image unsharpness and relative motion of the patient, the X-ray source and the detector must be minimized during the exposure. The third factor affecting resolution is related to the lateral spread of signal (light photons or electronic charges) in the detector from

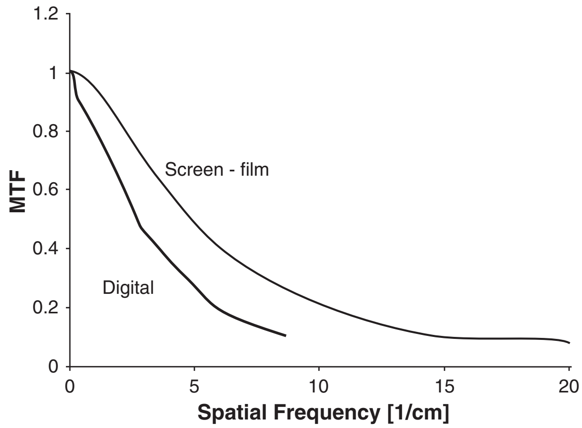

Fig. 1.7. MTF of a modern screen-film system (after Bunch 1997) and of a typical detector used in digital mammography

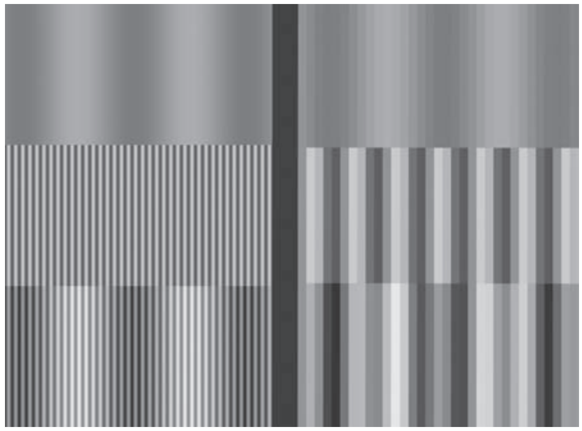

Fig. 1.8. Illustrating the concept of aliasing (from Pisano et al. 2004. With permission)

the point where an X-ray is absorbed to the point where the signal is recorded. Generally, as the detector is made thicker to increase η, more blurring of this sort will occur and therefore there is a compromise between spatial resolution and η. As will be discussed in Chap. 2 there are opportunities in the design of detectors for digital mammography to overcome, at least partially, the need to make this compromise.

In digital mammography systems, an additional factor, the spatial sampling, affects resolution. The X-ray signal from each del is averaged over the aperture, d. This causes the MTF of the detector to decrease. For practical purposes, detectors for digital mammography are not designed to provide as high a spatial resolution as obtainable from a film mammography system and as seen in Fig. 1.7 its MTF is typically considerably lower. Other factors such as contrast and noise characteristics provide offsetting improvements in imaging performance. Typically, the size of the del ranges from 50 to 100 µm (0.05–0.1 mm). If d is expressed in mm and the del is a square, then the MTF falls to 0 at a spatial frequency of $1/d$ cycles/mm in both the x and y directions. For example, a del with $d = 100$ µm, has an MTF that falls to 0 at 10 cycles/mm.

The spacing between dels or pitch, p is also an important factor. The larger that p is, the more information that will be lost through the sampling process and this restricts the maximum value of the spatial frequency of information in the image that can be represented accurately. The Nyquist theorem stipulates that for a pitch, p, the highest spatial frequency that can be accurately represented is:

$$f_N = \frac{1}{2p} \tag{1.13}$$

Higher spatial frequencies in the X-ray pattern will be mis-represented, a phenomenon known as *aliasing*. This is illustrated schematically in Fig. 1.8, where imaging of a pattern containing sinusoids with low spatial frequencies, high frequencies, and a mixture of the two are imaged with detectors having dels with fine and coarser pitch. Coarse sampling causes the erroneous creation of sinusoids of lower spatial frequency. So, undersampling both prevents reliable depiction of high spatial frequency information in the image, also because of interference due to these aliased low frequencies impairs the representation of the true low spatial frequency information in the image. According to (1.13), a detector with $d = 100$ µm and $p = 100$ µm will be susceptible to aliasing if there are spatial frequencies in the image above 5 cycles/ mm.

1.8 Detective Quantum Efficiency

The SNR or the SDNR are effective quantitative descriptors of the *quality* of the information carried by the radiological image. The larger the signal or signal difference is compared with the random fluctuation, the better the image is. As discussed above, SNR increases with increasing exposure and with higher values of η. It decreases when there are sources of noise other than quantum noise contributing to the image. The highest SNR occurs in the pattern of X-rays transmitted by the breast. This can be considered as the input signal. If the number of these X-ray quanta in a specified area was n_0, its value would be:

$$\mathrm{SNR_{in}} = \langle n_0 \rangle / \sqrt{\langle n_0 \rangle} = \sqrt{\langle n_0 \rangle}.$$

For a system that was perfect (i.e., no additional noise sources) except for incomplete absorption of all incident X-rays by the detector, the signal would be $\eta \langle n_0 \rangle$ and the noise $\sqrt{\eta \langle n_0 \rangle}$, giving a reduced SNR of $\sqrt{\eta \langle n_0 \rangle}$.

We can characterize the performance of the imaging system by determining how efficiently it transfers the input SNR to the system output (i.e., the observer). The *detective quantum efficiency* (DQE) computes the ratio:

$$\mathrm{DQE} = \frac{\mathrm{SNR}_{\mathrm{out}}^2}{\mathrm{SNR}_{\mathrm{in}}^2} \tag{1.14}$$

For a perfect system, DQE would equal 1.0. Considering just the efficiency of X-ray interaction described above, DQE would be: $\eta \langle n_0 \rangle / n_0 = \eta$, the fraction of incident X-rays used by the detector.

If there are other sources of noise, $\mathrm{SNR_{out}}$ will decrease below the value predicted by the number of interacting quanta so that DQE will fall below η. From the measurement of $\mathrm{SNR_{out}}$ it will appear that fewer X-rays have been used to form the image than has actually been the case and DQE is a measure of that apparent lack of efficiency. In fact, the quantity, $\mathrm{SNR^2}_{\mathrm{out}}$, which, in the absence of additional noise sources is just the number of X-rays detected, is known as the number of *noise-equivalent quanta* or NEQ.

It is common to present DQE and NEQ values as a function of spatial frequency (Bunch et al. 1987). DQE(f) specifies, at each level of detail, how well the system transfers the SNR information present at its input.

1.9 Energy Spectra for Digital Mammography

The X-rays emitted by the type of tube used in mammography are emitted over a range of energies, thus forming a spectrum. The shape of the energy distribution is controlled by the material forming the target of the X-ray tube, the type of metal foil used to filter the beam before it is incident on the patient, and the kilovoltage at which the tube is activated to make the exposure. In screen-film mammography, where there is little flexibility in controlling the aspect of the contrast provided by the film, the X-ray spectrum is chosen to provide the greatest practical contrast, C_{rad}. This tends to drive the choice toward relatively low energies, where the difference in attenuation coefficients between tissues is largest as was seen in Fig. 1.2, although the energy must be high enough to ensure that the breast is reasonably well penetrated. For an "average" breast, the examination is typically carried out at, e.g., 26 kV with a molybdenum target X-ray tube and a molybdenum filter placed in the beam as in Fig. 1.9. The use of a low energy causes the overall value of μ for breast tissue to be high, necessitating a relatively high radiation dose (1.10) to achieve an acceptable value of n_d (i.e., adequate SNR).

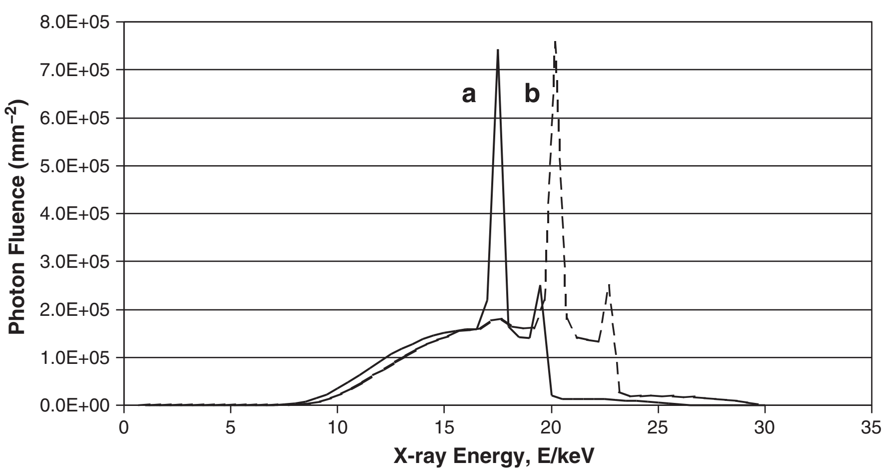

Fig. 1.9. X-ray spectra produced with (**a**) Mo target and Mo filter at 26 kV, (**b**) Rh target with Rh filter. With digital mammography the trend is toward use of more penetrating spectra

In digital mammography, obtaining an adequate value of SNR or SDNR is more important than achieving a specific value of C_{rad}, because the final contrast can be adjusted during image display. The SDNR does not change rapidly with X-ray energy and imaging with higher energies allows lower doses to be used and reduces the dynamic range requirements of the detector (Berns et al. 2003; Geertse et al. 2005; Lo et al. 2005; Bernhardt et al. 2006; Williams et al. 2008). Clinical practice in digital mammography is gradually shifting toward selection of higher kilovoltage and choice of rhodium rather than molybdenum filtration in Mo target systems or use of tubes equipped with rhodium or tungsten targets and appropriate filtration and kilovoltage to obtain a more penetrating beam than typically was used for screen-film mammography.

1.10 Clinical Dose Levels in Digital Mammography

The decoupling of image acquisition from display allows the brightness of the digital mammogram to be set at the desired level independently of the amount of X-ray exposure used to produce the image. Unlike the situation with film mammography, this provides great flexibility in choosing the dose for the examination. Nevertheless, there are important considerations in dose selection. Probably, the most important is to maintain an acceptable level of SNR or SDNR in the image. This implies that an adequate number of X-rays, n_d, are detected. In digital mammography, this can occur with a concommitent reduction in dose when compared with film if one or more of the following is true: (1) the DQE of the digital detector is higher than that of the screen-film image receptor, (2) the dose used for the film image is higher than that to achieve the required SNR, but is needed to produce a certain brightness and/or contrast, (3) a more penetrating X-ray beam can be used to produce the digital image so that less radiation is absorbed in the breast or (4) the scatter reduction mechanism for digital mammography is more efficient. In practice, for some digital mammography systems, dose reductions of between 25 and 30% compared with film mammography have been achieved. It is very important, however, that such reductions are obtained due to the above factors and not by degrading the SDNR at the risk of loss of diagnostic accuracy.

References

Barnes GT, Brezovich IA (1977) Contrast: effect of scattered radiation. In: Logan WW (ed) Breast carcinoma: the radiologist's expanded role. Wiley, New York, pp 73–81

Barnes GT, Brezovich IA (1978). The intensity of scattered radiation in mammography. Radiology 126:243–247

Barrett HH, Myers KJ (2001) Foundations of image science. Wiley, New York

Bernhardt P, Mertelmeier T, Hoheisel M (2006). X-ray spectrum optimization of full-field digital mammography: simulation and phantom study. Med Phys 33:4337–4349

Berns EA, Hendrick RE, Cutter GR (2003) Optimization of technique factors for a silicon diode array full-field digital mammography system and comparison to screen-film mammography with matched average glandular dose. Med Phys 30:334–340

Bunch PC (1997) The effects of reduced film granularity on mammographic image quality. In: Van Metter R, Beutel J (eds) Medical imaging 1997: physics of medical imaging. Proc. SPIE 3032, 302–317

Bunch PC, Huff KE, Van Metter R (1987) Analysis of the detective quantum efficiency of a radiographic screen-film combination. J Opt Soc Am A 4:902–909

De Almeida A, Rezentes PS, Barnes GT (1999). Mammography grid performance. Radiology 210:227–232

Geertse D, van Engen RE, Oostveen LJ, Thijssen MAO, Karssemeijer N (2005) Spectrum optimization for a selenium digital mammography system, Proceedings of the IWDM 2004, University of North Carolina, Chapel Hill, pp 116–122

Haus AG, Yaffe MJ (2000) Screen-film and digital mammography. Radiol Clin North Am 38(4):871–898

ICRU (2005) International Commission on Radiological Units. *Patient Dosimetry for X Rays Used in Medical Imaging*, ICRU Report 74, *JICRU* **5**, Oxford University Press, Oxford, UK

Johns, PC, Yaffe, MJ (1987). X-ray characterization of normal and neoplastic breast tissues. Phys Med Biol 32:675–695

Lo JY, Samei E, Jesneck JL, Dobbins JT III, Baker JA, Singh S, Saunders RS, Floyd CE (2005) Radiographic technique optimization for an amorphous selenium FFDM system: Phantom measurements and initial patient results, Proceedings of the IWDM 2004, University of North Carolina at Chapel Hill,, pp 31–36

Wagner AJ (1991) Contrast and grid performance in mammography. In: Barnes GT, Frey GD (eds) Screen film mammography: imaging considerations and medical physics responsibilities Medical Physics Publishing, Madison, WI, pp 115–134

Williams MB, Raghunathan P, More MJ, Seibert JA, Kwan A, Lo J, Samei E, Ranger NT, Fajardo L, McGruder A, McGruder S, Maidment A, Yaffe MJ, Bloomquist A, Mawdsley G (2008) Optimization of exposure parameters in full field digital mammography. Med Phys 35(6):2414–2423

Pisano ED, Yaffe MJ, Kuzmiac CM (2004) Digital mammography. Lippincott, Williams & Wilkins, Philadelphia

Detectors for Digital Mammography

2

Martin J. Yaffe

CONTENTS

M. J. Yaffe, PhD
Imaging Research Program, Sunnybrook Health Sciences Centre, University of Toronto, 2075 Bayview Avenue, Toronto, ON, Canada M4N 3M5

KEY POINTS

The X-ray detector is the heart of a digital mammography system. Its improved characteristics of dynamic range and signal-to-noise ratio provide inherent advantages over screen-film technology. Detector technologies used for digital mammography can be distinguished by the acquisition geometry into scanning or full-field detectors, by energy conversion mechanism into phosphor-based and nonphosphor-based detectors and by how the detector signal is converted into an image value into signal-integrating and quantum-counting systems. Reading of the detector signal can be integrated into the detector assembly or the detector can be in the form of a sensitive plate in a portable cassette, which is moved to a separate device for readout. Detector performance characteristics vary among these different technological approaches. An understanding of the physics on which detector operation is based can help explain these differences. Various image processing operations can be carried out to correct for spatial nonuniformities in detector response and to improve the effective spatial resolution of the detector. In addition, use of a digital detector provides opportunities for more sophisticated automatic control of exposure factors for image acquisition.

2.1 Introduction

The detector is one of the defining features of a digital mammography system. The detector produces an electronic signal that represents the spatial pattern of

X-rays transmitted by the breast. The detector is designed to overcome several of the limitations inherent in the screen-film image receptor used in analog mammography, and in so doing, potentially provides improved diagnostic image quality and a reduction of dose to the breast.

The function of the detector can be described by a set of sequential operations that include:

(a) Interaction with the X-rays transmitted by the breast and absorption of the energy carried by the X-rays
(b) Conversion of this energy to a usable signal – generally light or electronic charge
(c) Collection of this signal
(d) Conversion of light to electronic charge (in the case of phosphor-based detectors)
(e) Readout of charge, amplification, and digitization

These operations must be optimized if the detector is to provide high-quality images at appropriate dose levels. Detectors are characterized by their quantum detection efficiency, sensitivity, spatial resolution properties, noise, dynamic range, and linearity of response.

As discussed in Chap. 1, digital images are acquired by sampling the pattern of X-rays transmitted by the breast. In practice, this is often accomplished using a detector that is constructed as an array of discrete detector elements or *dels*, each of which more or less independently measures the X-rays incident on it. The pitch or spacing between dels and the dimensions of the active portion of each del (aperture) in part determine the spatial resolution properties of the imaging system. The concept of spatial resolution and its quantification in terms of the modulation transfer function (MTF) were introduced in Chap. 1.

2.2 Geometric Considerations

A detector that is suitable for mammography must be able to capture the transmitted X-ray pattern from as much of the breast as possible. To satisfy this requirement, it must have spatial dimensions of approximately 24 × 30 cm. Positioning of a small breast can be facilitated if a smaller detector size (format of approximately 18 × 24 cm) is also available. Alternatively, the imaging system can be designed such that when a small breast is imaged on 24 × 30 cm detector, the collimated X-ray beam and the detector area over which the image is recorded can be shifted to cover the appropriate tissue in the axilla, depending on whether the left or right breast is being imaged.

It is important to be able to image the breast as close to the chest wall as possible. Therefore, there should be the minimum possible amount of insensitive material associated with the detector at this edge of the image. Finally, an excessively thick detector assembly can impede positioning of the breast for some views.

Two major types of digital mammography systems have been introduced. One uses a detector that is the full size of the field that is to be imaged. The other employs a detector array that is long in one dimension and narrow in the other. In the first type, "snapshot" imaging is performed by acquiring the X-ray transmission information from all parts of the breast simultaneously. As in film-based imaging, a radiographic grid is generally used in such systems to reduce the loss of quality caused by recording scattered X-rays.

In the second type of imager, the detector is scanned across the breast in synchrony with a long, narrow collimated X-ray beam to acquire the image progressively. Because the X-ray field is smaller, there is less scattered radiation recorded and these systems are operated without a radiographic grid.

2.3 Basic Physics of X-Ray Detectors

The initial stage in the detector, X-ray interaction is common to all detector technologies. This occurs at the level of individual atoms of the detector material. At the energies used for mammography, X-rays incident on the detector interact by one of the three mechanisms, elastic scattering, inelastic (Compton) scattering, or the photoelectric effect. Elastic scattering leaves no energy in the detector and produces no signal. Compton scattering results in part of the energy of the incoming X-ray being absorbed at the initial point of X-ray impact liberating an energetic recoil electron, but the remainder is carried away by the scattered quantum to be deposited elsewhere, resulting in the loss of spatial resolution. In a photoelectric interaction, the incoming X-ray knocks an electron out of an inner (K-shell or L-shell) orbital of the atom (Fig. 2.1) and much of the energy of the X-ray is transferred to this "photoelectron." When the vacancy is refilled by an electron from a more loosely bound shell, the remainder of the energy is