Digital Radiography

Physical Principles and Quality Control

Euclid Seeram

Second Edition

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ISBN 978-981-13-3243-2 ISBN 978-981-13-3244-9 (eBook)
https://doi.org/10.1007/978-981-13-3244-9

Library of Congress Control Number: 2018965417

This Springer imprint is published by the registered company Springer Nature Singapore Pte Ltd.
The registered company address is: 152 Beach Road, #21-01/04 Gateway East, Singapore 189721, Singapore

This book is dedicated to my two smart, precious, and overall cute granddaughters, Claire and Charlotte, with love and blessings to you both forever.

Foreword

Planar or general radiography is the most common imaging examination in medicine. It is an important tool for clinicians to consider when making a diagnosis of an illness or injury for a patient.

Digital radiography is now the method to capture and display radiographic images. Digital radiography is a modality within the highly technological imaging world of computed tomography, magnetic resonance imaging, diagnostic ultrasound, positron emission tomography, and other complex digital imaging modalities. Digital radiographs require complex image processing to enable clinicians to fully visualize image detail. Digital images must be storable, transferrable, and recallable at distant locations and must meet quality standards to provide the required diagnostic patient information.

Radiographers/radiologic technologists and radiologists must understand the physical principles and technical details of digital radiography imaging systems such as computed radiography, flat-panel digital systems, digital fluoroscopy, and digital mammography in order to maximize image quality from the captured X-ray data. This text provides an insight into the understanding of digital radiography imaging modalities. It describes the acquisition of these images in the context of the digital environment through an explanation of the basic digital image processing, image storage, and transmission and explains the process of quality control in digital radiography.

The name Euclid Seeram has become synonymous with radiography education. Euclid has decades of experience in teaching all the areas of medical imaging. During this time he has gained worldwide respect as an educator and an author. Euclid has the gift of being able to explain difficult concepts in a way that students can grasp. This has been seen in his other texts. This book continues in this tradition and makes complex concepts easier to understand.

The success of the first edition is only part of the story of Euclid's contributions in the field. He continues to promote updated educational content for technologists, including a website of DR artifacts. By the time I met Euclid again at one of a series of CR and DR workshops that Larry Filipow, a Canadian pioneer in DR and PACS, Tony Seibert, an American pioneer in DR and PACS, and I were presenting across Canada, his work was well known throughout North America. Coincidentally, in individual conversations with Euclid, each of us said essentially the same thing: "we like what you're doing for the technologists!"

Through his efforts to provide practical and understandable information to the individuals who have the responsibility to produce the DR images, Euclid has made a broad positive impact on the quality of DR operations.

Students will appreciate the detail, ease of explanation, and breadth of treatment of the topics in digital radiography. His mother properly named him Euclid expecting him to complement this moniker for science and mathematics. Furthermore, Euclid has demonstrated a continual quest for knowledge in the medical imaging sciences field via his numerous lifelong education pursuits. Having completed courses in digital radiography from several notable experts in the field as well as a course on medical imaging informatics from Stanford University, he has also obtained a certificate of MRI Physics from the American Association of Physicists in Medicine summer school. He is a full member of the Health Physics Society and has been active in the field of Radiation Protection, which led to his PhD dissertation entitled "*Optimization of the Exposure Indicator of a Computed Radiography Imaging System as a Radiation Dose Management Strategy.*"

Euclid is willing to share this knowledge with the imaging community. These efforts continue to make him a successful author and educator. For those studying in the field of medical imaging or for those just wishing to gain a high level understanding of digital radiography, we highly recommend this book.

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Preface

The motivation for a second edition of this book stems from the continued *technical evolution* of digital radiographic imaging systems, as well as the introduction of new principles, issues, and ideas relating to digital imaging. With these considerations in mind, the book is now titled *Digital Radiography: Physical Principles and Quality Control.* Furthermore, technical advances have been a recurring feature in digital detector technology and the development of a standardized exposure indicator by the International Electrotechnical Commission (IEC), Picture Archiving and Communication Systems (PACS), and imaging informatics. The next era in imaging informatics will include topics such as enterprise imaging, cloud computing, big data, and artificial intelligence (AI) including subsets of AI such as machine learning and deep learning.

Furthermore, there has been a significant development of a technique referred to as digital tomosynthesis featuring two main application areas: radiographic tomosynthesis and digital breast tomosynthesis (DBT).

An important consideration from this new title is the lack of textbooks on this subject for radiologic technologists especially one that includes dose optimization principles, an essential ingredient that provides significant guiding principles to clinical practitioners to work within the ALARA (as low as reasonably achievable) philosophy, in an effort to protect not only the patient but personnel as well.

New to This Edition

The chapters in this book have been updated to address the continued technical evolution and major technology trends in digital radiographic imaging. Furthermore, a set of updated articles have been cited to support the inclusion of major new technical areas of current interest in clinical practice such as the following: the standardized exposure indicator, digital tomosynthesis, and enterprise imaging, including vendor neutral archives (VNAs), cloud computing, big data (BG), and artificial intelligence (AI) including subsets of AI such as machine learning and deep learning. The addition of numerous new images illustrates the fundamental principles of these new technologies, and new line drawings serve to enhance and provide important concepts in the text. Last but not least, a reasonable effort has been

made to keep the cited literature current. These references are important since they serve a twofold purpose:

1. To validate the statements made in the textbook regarding digital radiography physical principles and applications
2. To guide the student to the primary and secondary sources of information that serve as the fundamental basis for pursuing their own research and presentations

Purpose

The purpose of this textbook is fourfold as follows:

1. To provide a comprehensive coverage of the physical principles of digital radiography imaging systems and associated technologies, such as Picture Archiving and Communication Systems (PACS), VNAs, and new topics in medical imaging informatics, such as cloud computing, VNAs, BG, digital tomosynthesis, and AI and its subsets
2. To lay the theoretical foundations necessary for the effective use of digital radiography in clinical practice
3. To enhance communication among radiology personnel such as technologists, radiologists, medical physicists, biomedical engineers, and between radiology personnel and vendors
4. To meet the educational requirements of various radiologic technology professional organizations including the American Society of Radiologic Technologists, the American Registry of Radiologic Technologists, the Canadian Association of Medical Radiation Technologists, the College of Radiographers in the United Kingdom, as well as those in Africa, Asia, Australia, and continental Europe

Content and Organization

The content and organization of the book have not been changed significantly; however, certain chapters have been deleted completely, and some content has been reshaped and added to "new" chapters. For example, in the first edition, *Chap. 4*, *Effective Use of Computed Radiography (CR)*, has been deleted. This chapter was written by a dear friend of mine Barry Burns, MS, RT (R), DABR. Barry was a Professor—Division of Radiologic Science, School of Medicine, University of North Carolina, Chapel Hill, North Carolina. Barry passed away in 2016 (God bless him).

Another chapter, Chap. 10, *Quality Control for Digital Radiography*, that was part of the first edition, was written by Dr. Charles Willis, PhD, DABR, Associate Professor, Department of Imaging Physics, University of Texas, MD Anderson Cancer Center, Houston, Texas. In this chapter, several topic areas have been included, in particular Sections Sect. 11.4 "Understanding Processes and Errors in Digital Radiography" and Sect. 11.7 "Responsibilities

for DR QC." Such materials are attributed to Dr. Charles Willis, PhD. I am grateful to Dr. Willis, who graciously gave his permission to reuse these materials. This chapter is Chap. 11 in this book, and the new title is *Continuous Quality Improvement for Digital Radiography.* Furthermore *three new chapters have been added, Chaps. 5, 8, and 12*.

The content and organization of the book are based on the following structures:

- Brief historical developments of digital radiography imaging systems
- Digital image processing fundamentals
- Physical principles and technological aspects of digital radiography imaging modalities
- Effective use of technologies that are integral components of digital imaging systems, for example, tomosynthesis including DBT, PACS, imaging informatics, and quality control procedures

Chapter 1 presents an *Overview of the Technologies* that constitute the subject matter of this book including a discussion of the limitations of film-based radiography and the major structural components of digital radiography imaging systems as well as brief description of the image acquisition detectors, PACS, and imaging informatics.

Chapter 2 provides a detailed description of the topic of *Digital Image Processing*, a topic that is of particular importance in today's digital clinical imaging practice.

Chapter 3 deals with the basic physics and technology of *Computed Radiography* (CR) and describes, specifically, the physics of photostimulable luminescence, CR technical components, image processing, exposure control in CR, and image quality descriptors for CR and presents an overview of artifacts. Finally the chapter concludes with a brief introduction of selected quality control tests for CR, using quality criteria for assessment.

Chapter 4 is devoted to *Flat-Panel Digital Radiography* imaging systems and includes a description of the different types of flat-panel digital imaging systems, design characteristics, operating principles, image processing, and imaging performance characteristics such as spatial resolution, modulation transfer function, dynamic range, detective quantum efficiency, image lag, and image artifacts. Wireless digital detectors are briefly introduced.

Chapter 5 is a new chapter focusing on details of the *Standardized Exposure Indicator* (EI) through a description of the conditions for the IEC standardized EI, various definitions, determination of the standardized EI, interpretation of deviation index (DI) values, factors affecting EI values, responsibilities, and an objective approach for establishing target EI values.

Chapter 6 provides a description of the technical aspects and image processing for *Digital Fluoroscopy Systems*, based on image intensifiers and flat-panel digital detectors, as well as the fundamental physical principles of digital subtraction angiography.

Chapter 7 describes the essential technological aspects of *digital mammography*. Topics include types of detectors, digital image processing, and

various applications such as digital tomosynthesis and computer-aided detection and diagnosis.

Chapter 8 is a *new chapter* dealing *Digital Tomosynthesis*. Topics include definition and principles (image acquisition and system components, image acquisition parameters, image reconstruction, image display and communication), radiation dose considerations, synthesized 2D digital mammography, and clinical applications.

Chapter 9 examines the major components, and *core technologies of PACS* are described in detail followed by an outline of the bare essentials of information systems and communication standards for digital radiology, including DICOM (Digital Imaging and Communications in Medicine), HL-7 (Health Level-7), and the technical framework of IHE (Integrating the Healthcare Enterprise). Furthermore the concepts relating to vendor neutral archives (VNAs) and enterprise imaging (EI), including a definition of EI and the major elements of an enterprise imaging system, are presented.

Chapter 10 outlines the major components of medical *imaging informatics*, an evolving field for the digital radiography community. Examples of these components include the health information systems, electronic health record, systems integration, information technology (IT) security fundamentals, and skills and certification of a PACS technologist. Additionally *new topics* that are considered emerging topics in imaging informatics are introduced in this chapter. In particular, these topics include *cloud computing*, *big data*, *artificial intelligence*, *machine learning*, and *deep learning*.

Chapter 11 presents a detailed description of the elements of *Continuous Quality Improvement in Digital Radiography*. Major topics described include significant elements of continuous quality improvement (CQI) such as definitions of quality assurance (QA) and quality control (QC), dose optimization, parameters for QC monitoring in digital radiography, and tolerance limits or acceptance criteria. The next important section of this chapter reviews the processes and errors in digital radiography, responsibilities for digital radiography QC, and an overview of digital mammography QC (topics that were addressed in detail in the first edition of this book and written by Dr. Charles Willis, PhD, as mentioned above, under the topics of "Content and Organization"). In addition, four QC tests for digital radiography have been described in a manner that students can conduct them in a radiography laboratory.

Chapter 12 is a *new chapter* and the final chapter in this book. It deals primarily with *Dose Optimization in Digital Radiography*, through a description of topics such as dose optimization approaches (optimization of kVp, optimization of mAs, optimization of the exposure indicator (EI)) and dose optimization tools and methods for image quality assessment. Finally the chapter concludes with an overview of the method of visual grading of normal anatomy and optimization of the EI as a dose management strategy.

Use and Scope

The purpose of this comprehensive text is to meet the wide and varied requirements of its users, students, and educators alike. Therefore this book can meet many different educational and program needs. *Digital Radiography: Physical Principles and Quality Control* can be used as the primary text for introductory digital imaging courses at the diploma, associate, and baccalaureate degree levels. Additionally, it can be used as a resource for continuing education programs; it functions as a reference text for others working in radiology, for example, biomedical engineering professionals. Finally it provides the required overview of the physical principles and technological considerations and may be viewed as providing the needed prerequisites for graduate-level (master's degree) courses in digital radiography.

The content is intended to meet the educational requirements of various radiologic technology professional associations including the *American Society of Radiologic Technologists* (ASRT), the *American Registry of Radiologic Technologists* (ARRT), the *Canadian Association of Medical Radiation Technologists* (CAMRT), the *College of Radiographers* in the United Kingdom, as well as those professional medical imaging organizations in Africa, Asia, Australia, and continental Europe.

Digital radiography has become an integral part of the education of radiologic technologists and related professionals who play a significant role in the care and management of patients undergoing both routine and other sophisticated imaging procedures.

Enjoy the pages that follow, and remember that with your knowledge and skills in digital radiography, your patients will benefit from your wisdom.

British Columbia, CANADA Euclid Seeram

Acknowledgments

The single most important and satisfying task in writing a book of this nature is to acknowledge the help and encouragement of those individuals who perceive the value of its contribution to the medical imaging sciences literature. It is indeed a pleasure to express sincere thanks to several individuals whose time and efforts have contributed tremendously to this third edition.

First and foremost, I am indebted to Dr. Charles Willis, PhD, Department of Imaging Physics, University of Texas, MD Anderson Cancer Center, Houston, Texas, who provided his support for me to reuse a section of his chapter in the first edition of this book. Thanks Chuck. Indeed I have learned a great deal from you attending your workshops and through your educative published papers in the literature.

Furthermore I am grateful to my good friend and colleague Anthony Chan, MEng, MSc, PEng, CEng, CCE, at the British Columbia Institute of Technology (BCIT), a Canadian award-winning Biomedical Engineer, who provided good explanations of the engineering aspects of making digital detectors and other technical aspects of PACS, such as DICOM and HL-7. In addition, I must also thank Bruno Jaggi, DiplT, BASc, MASc, PEng, an expert Biomedical Engineer in the Faculty of Applied Sciences at the University of British Columbia. Bruno has always been supportive in teaching me about the Fourier transform and its applications in medical imaging. Additionally his course and regular discussions on digital image processing provided me with the theoretical background for a better understanding of image post-processing operations.

The content of this book is built around the works and expertise of several noted medical physicists, radiologists, computer scientists, and biomedical engineers who have done the original research. In reality, they are the tacit authors of this text, and I am truly grateful to all of them. In this regard, I owe a good deal of thanks to Dr. Anthony Siebert, PhD, of the University of California at Davis and Dr. Charles Willis, PhD, of the University of Texas, MD Anderson Cancer Center, two expert physicists in digital radiography and from whom I have learned the physics and technical aspects of digital radiography through their seminars and workshops that I have attended. Furthermore I am also grateful to several other physicists from whom I have learned much about digital imaging physics through their published writings. These include Dr. Perry Sprawls, PhD (Emory University); Dr. Kerry Krugh, PhD (Medical Physicist, University of Toledo Medical Center, Toledo, Ohio); Dr. John Yorkston, PhD (Senior Research Scientist, Clinical Applications

Research, Carestream Health Inc., Rochester, NY); Dr. Martin Spahn, PhD (Siemens AG, Medical Solutions); Dr. Ehsan Samei, PhD (Medical Physicist and Associate Professor at Duke University); Dr. Beth Schueler, PhD (Medical Physicist, Mayo Clinic); and Jeff Shepard, MS (Senior Medical Physicist, MD Anderson Cancer Center). One more medical physicist to whom I owe thanks is Dr. John Aldrich, PhD, Vancouver General Hospital, University of British Columbia, whose seminars on digital radiography and other topics have taught me quite a bit. Thanks John. Furthermore, I owe a good deal of thanks to my good friend and colleague Dr. Rob Davidson, PhD, MAppSc (MI), BBus, FIR, Professor of Medical Imaging, University of Canberra, Australia. Rob's advice on the inclusion of digital breast tomosynthesis was much appreciated.

I must acknowledge the efforts of all the individuals from digital radiography vendors who have assisted me generously with technical details and photographs of their systems for use in the book. In particular, I am grateful to Joanne Muldoon of Hologic® and Senior Scientist Dr. Andrew Smith, PhD, at Hologic®. Thanks so much for this assistance. Other vendors are acknowledged in respective figure legends.

In this book I have used several illustrations and quotes from original papers published in the professional literature, and I am indeed thankful to all the publishers and the authors who have done the original work and have provided me with permission to reproduce them in this textbook. I have purposefully used several quotes so as not to detract from the authors' original meaning. I personally believe that these quotes and illustrations have added significantly to the clarity of the explanations. In this regard, I am appreciative of the Radiological Society of North America (RSNA) and Jamie Dulkowski, Manager of Informatics at the RSNA; *Academic Radiology*; Wolters Kluwer Health; the American Association of Physicists in Medicine (AAPM), for materials from *Medical Physics*; Wiley-Blackwell Publishers Inc.; and Manning Publications Co., Shelter Island, NY.

The people at Springer Nature, SPi Global, deserve special thanks for their hard work, encouragement, and support of this project. They are Dr. Naren Aggarwal, Executive Editor of *Clinical Medicine*, who was very supportive of the initial proposal for this textbook and Ms. Abha Krishnan, Project Coordinator (Books), at Springer Nature, who provided support and maintained continued communication to bring this work to fruition. I must also thank the individuals in the production department at Springer for doing a wonderful job on the manuscript to bring it to its final form. In particular, I am grateful to members of the production team, who have worked exceptionally hard during the production of this book, especially in the page-proof stage.

Finally, my family deserves special mention for their love, support, and encouragement, while I worked into the evenings on the manuscript. I appreciate the efforts of my lovely wife, Trish, a warm, caring, and overall special person in my life; our son David, a brilliant young editor and publisher of his own online digital photography magazine called *PhotographyBB* (available at Photographybb.com) that has a worldwide audience; and our beautiful daughter-in-law, Priscilla, a smart, remarkable, and caring young woman.

Thanks for taking such excellent care of our two granddaughters, to whom this book is dedicated. I love you all.

Last but not least, I must thank my students who have diligently completed my courses on digital imaging modalities, PACS, and digital image processing in radiology, at both the diploma and degree levels. Thanks for all the challenging questions.

Keep on learning and enjoy the pages that follow.

British Columbia, CANADA Euclid Seeram

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Digital Radiography: An Overview

1

Abstract

This chapter provides a broad overview of the elements of digital radiography (DR). First a definition of digital radiography is presented followed by a brief review of the essential underlying principles of film-screen radiography (F-SR) including its limitations. Secondly, the major components of digital radiography imaging systems are briefly outlined. The third topic covered a comprehensive explanation of digital imaging modalities such as computed radiography (CR), flat-panel digital radiography (FPDR), digital fluoroscopy, and digital mammography. Furthermore, the elements of Picture Archiving and Communication Systems and quality control are introduced. Finally, the chapter concludes with an introduction to topics such as enterprise imaging, cloud computing, Big Data, machine learning, and deep learning as well as artificial intelligence that are considered the next era of imaging informatics.

1.1 Introduction

Film-screen radiographic imaging has been the workhorse of radiology ever since the discovery of X-ray by WC Roentgen in 1895. Today film-screen radiography is now obsolete and has been replaced by *digital radiographic imaging* or simply *digital radiography*. Digital radiography technologies include not only digital image acquisition modalities but also digital image processing, display, storage, and image communication. Specifically, digital image acquisition modalities include computed radiography, flat-panel digital radiography, digital mammography, and digital fluoroscopy for routine gastrointestinal fluoroscopy and vascular imaging. Furthermore the other technologies of image processing, display, storage, and image communication using computers fall within the domain of *information technology*. The application of information technology to digital radiography has created an area of study and practice popularly referred to as *medical imaging informatics* [1, 2].

The next era in medical imaging informatics includes at least six major topics that have received increasing attention in the literature, and which are important for all those working in medical imaging and radiation sciences. These include enterprise imaging, Big Data, cloud computing, machine learning and deep learning, and artificial intelligence.

The purpose of this chapter is to present a broad overview of the elements of digital radiography imaging systems including medical imaging informatics and to lay the overall framework and foundations for the rest of this book.

E. Seeram, *Digital Radiography*, https://doi.org/10.1007/978-981-13-3244-9_1

1.2 Digital Radiography: A Definition

The American Association of Physicists in Medicine (AAPM) has offered a definition of digital radiography as "radiographic imaging technology producing digital projection images such as those using photostimulable storage phosphor (computed radiography, or CR), amorphous selenium, amorphous silicon, charge-coupled device (CCD), or metal oxide semiconductor-field effect transistor (MOSFET) technology" [3]. These technologies can produce acceptable image quality over a wider range of exposure techniques compared to film-screen radiography.

Digital detectors capture and convert X-ray attenuation data from the patient into electronic signals (analog signals) that are subsequently converted into digital data for processing by a digital computer. The result of processing is a digital image that must be converted into one that can be displayed on a computer monitor for viewing by an observer. The major system components and the essential steps to digital image production are illustrated in Fig. 1.1. The displayed image can be manipulated using a variety of digital image processing techniques to enhance the interpretation of diagnostic radiology images [4, 5].

1.3 Film-Based Radiography: A Brief Review

1.3.1 Basic Steps in the Production of a Radiograph

The production of a film-based radiographic image involves several steps as illustrated in Fig. 1.2. X-rays pass through the patient and fall upon the film to form a latent image. The latent image is then rendered visible using *chemical processing* and subsequently displayed on a light viewbox for viewing and interpretation by a radiologist. The film image appears with varying degrees of blackening as a result of the amount of exposure transmitted by different parts of the anatomy. While more exposure produces more blackening, less exposure produces less blackening, as is clearly illustrated in Fig. 1.3. Additionally, while the blackening is referred to

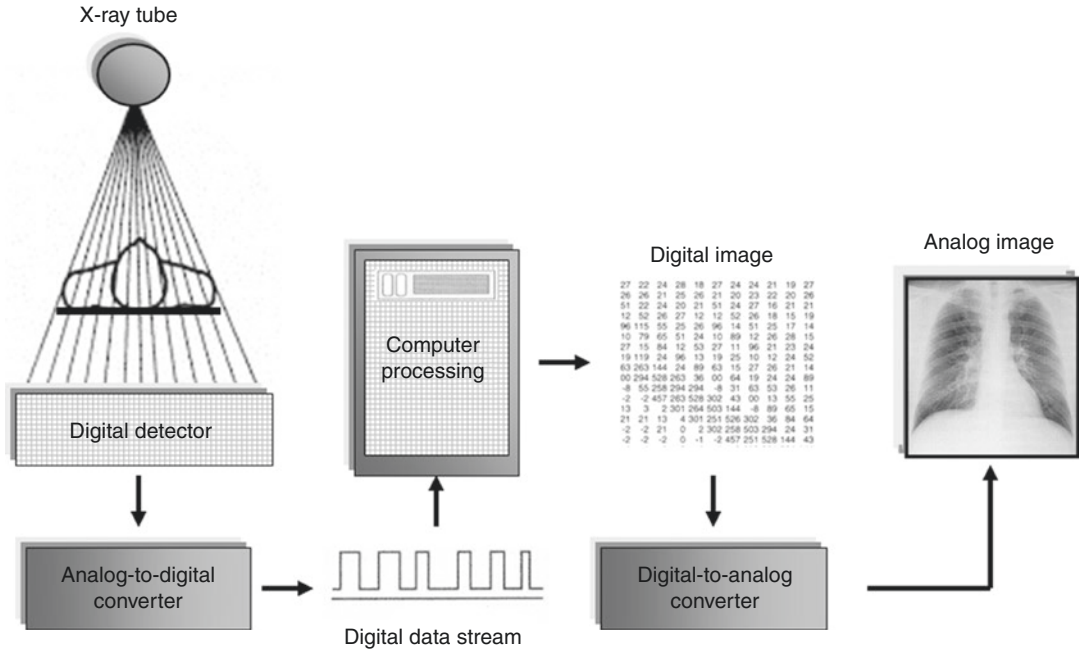

Fig. 1.1 The major system components and the essential steps in the production of a digital image

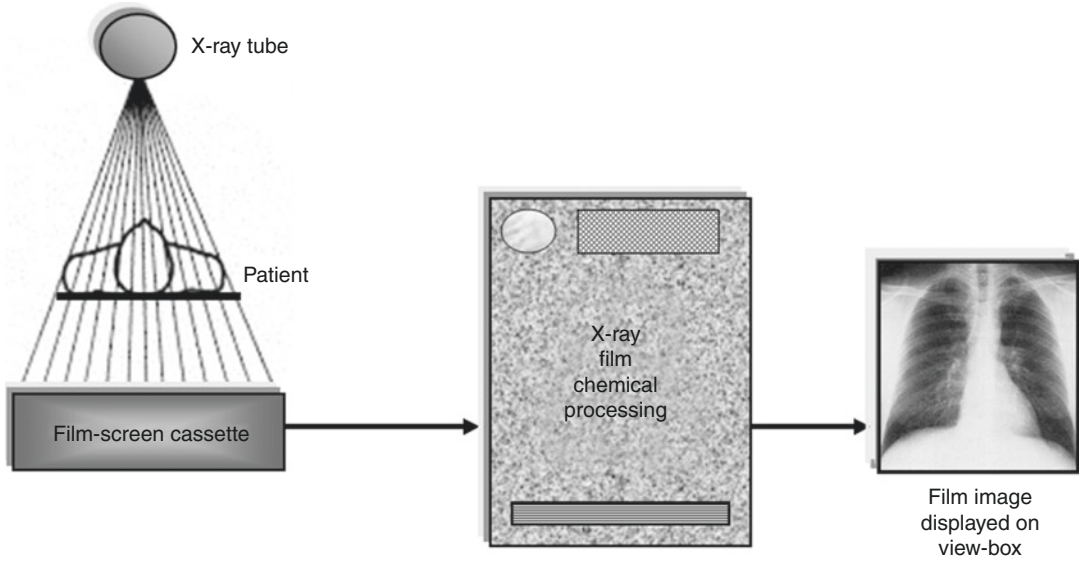

Fig. 1.2 The basic steps in the production of a film-based radiographic image

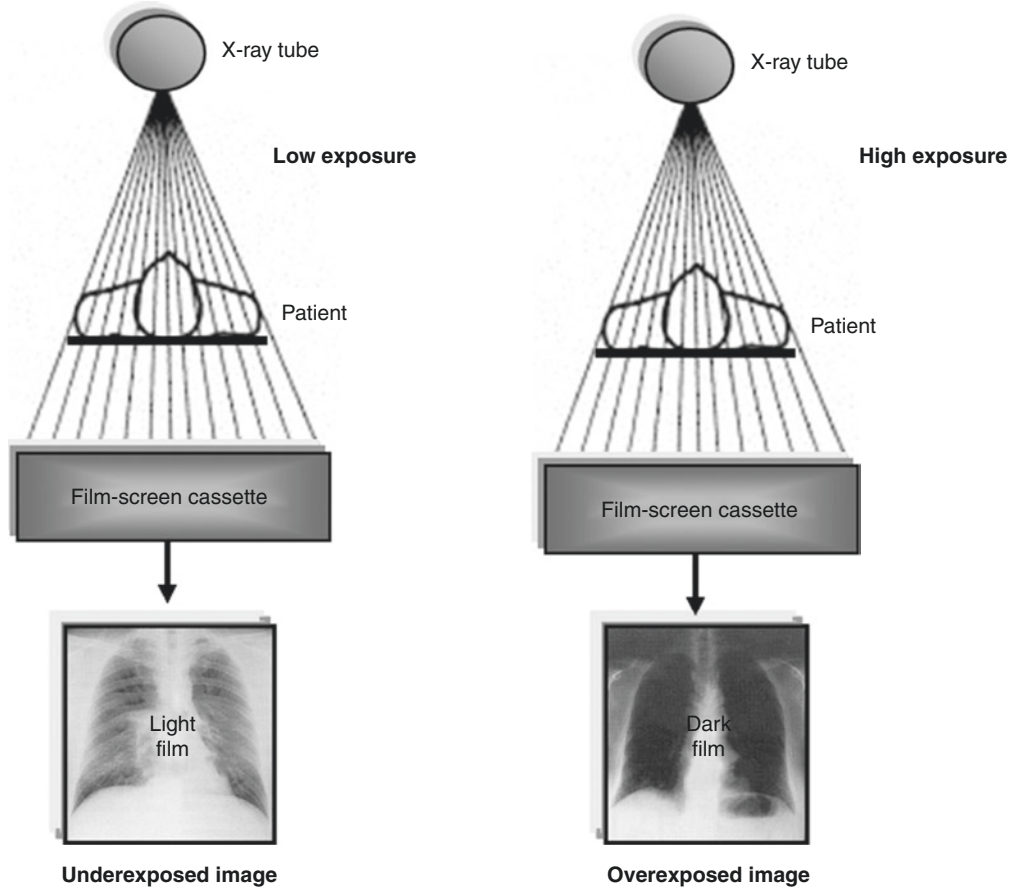

Fig. 1.3 The visual image quality feedback in film-based radiography as a result of low and high exposures

as the film density, the difference in densities in the image is referred to as the film contrast. The film therefore converts the radiation transmitted by the various types of tissues (tissue contrast) into film contrast.

An image displayed on a light viewbox transmits light into the eyes of the radiologist who interprets the image. This transmitted light can be measured using a densitometer and is referred to as the *optical density* (OD), which is defined as the log of the ratio of the intensity of the viewbox (original intensity) to the intensity of the transmitted light. The OD is used to describe the degree of film blackening as a result of radiation exposure.

1.3.2 The Film Characteristic Curve

The film contrast can be described by what is popularly known as the film *characteristic curve* or the Hurter-Driffield (H and D) curve. The curve is a plot of the optical density (OD) to the radiation exposure (or more accurately, the logarithm of the relative exposure) used in the imaging process. The purpose of the curve is to indicate the degree of contrast or different densities that a film can display using a range of exposures. An idealized characteristic curve for film-screen radiography, which is shown in Fig. 1.4, has three main segments, the toe, the slope (straight line portion), and the shoulder. While the toe and shoulder indicate underexposure and overexposure, respectively, the slope represents the useful portion of the curve and reflects acceptable exposure or the range of useful densities. This simply means that if an exposure falls at the toe region (OD = 0.12–0.20), the image will be light and generally useless. If the exposure falls in the shoulder region of the curve (OD = about 3.2), the image will be black and serve no useful purpose in providing a diagnosis. If the exposure falls within the slope of the curve (OD = 0.3–2.2), the image contrast (density) will be acceptable, and this region of the curve contains the useful range of exposures.

There are four other factors that can be described using the characteristic curve. These include the film speed, average gradient, the film gamma, and the film latitude. Only film speed and film latitude will be reviewed here. The

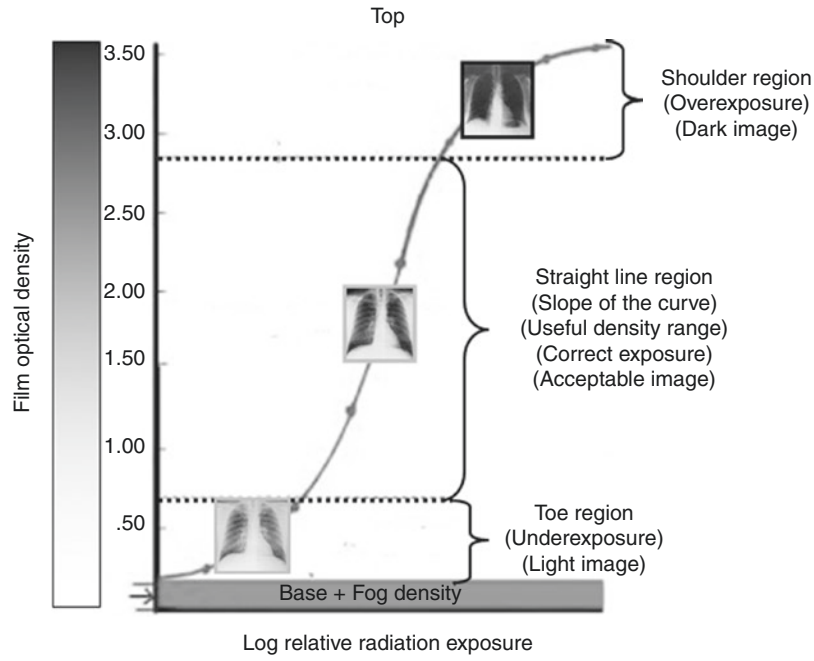

Fig. 1.4 The idealized characteristic curve (H and D curve) for film-screen radiography. See text for further explanation

interested reader should refer to any standard radiography physics text for a further description of the other terms.

The *film speed* refers to the sensitivity of the film to radiation, and it is inversely proportional to the exposure (E) and can be expressed algebraically as:

$$\text{Film speed} = 1 / E$$

This means that films with high speeds (fast films) require less exposure than films with low speeds (slow films). The film latitude on the other hand describes the range of exposures that would produce useful densities (contrast). While wide exposure latitude films can respond to a wide range of exposures, films with narrow exposure latitude can respond only to a small range of exposures. In the latter situation, the technologist has to be very precise in the selection of the exposure technique factors for the anatomical part under investigation. This is mainly because the film emulsion has a nonlinear response to chemical processing [6].

1.3.3 Exposure Technique Factors

Exposure technique factors affect the quantity and quality of the X-ray beam coming from the X-ray tube during the examination. These factors are set by the technologist on the console and are specifically selected for the patient size and type of examination and include the kilovolt peak (kVp), the tube current or milliamperage (mA), and the exposure time in seconds (s). While the kV controls the X-ray beam quality or the penetrating power of the X-ray beam, the mA controls the X-ray beam quantity that is the number of photons in the beam. The exposure time on the other hand determines the length of time that the X-ray beam is on during the exposure to the patient. The combination of mA and s results in the mAs which controls the X-ray beam quantity, in the same manner as mA and s used separately.

Exposure technique factors affect the contrast and density of the images on film. This will be described briefly below.

1.3.4 Automatic Exposure Control

The use of manual selection of exposure technique factors sometimes results in poor exposures that ultimately affected the image on the film. The problem is solved with the use of automatic exposure control (AEC) in which a preset quantity of radiation reaching the film image detector (cassette) is automatically measured and terminated when a preset optical density (degree of blackening, in the case of film) or signal-to-noise ratio (SNR) in the case of a digital detector is reached [6].

1.3.5 Image Quality Factors

The overall goal of radiographic imaging is to produce optimum image quality. The characteristics of a film that determine its quality are resolution, contrast, noise, distortion, and artifacts. Only the first three will be briefly described here.

Resolution refers to a characteristic that allows an observer to see separate objects on a film. There are two types of resolution, spatial resolution and contrast resolution. While the former refers to detail or sharpness (or visibility of detail), the latter refers to the differences in tissue contrast that can be seen on the film. *Spatial resolution* refers to detail, and it is measured in line pairs per millimeter (lp/mm). The greater the lp/mm a film can demonstrate, the greater the detail that can be seen. Detail is affected by several factors such as the focal spot size, motion of the patient, and the image receptor design characteristics. Detail is optimum when small focal spots are used, when the patient does not move during the exposure, and when detail cassettes are used.

Contrast is a significant image quality characteristic. It is the density differences on a radiographic image. A high-contrast image is characterized by regions of high density (dark) and low density (light). Several factors affect the contrast of a radiographic image, including the object, energy of the beam, scattered radiation, grids, and the film.

The main controlling factor for image contrast, however, is kVp. Optimum contrast is produced when low kVp techniques are used. A grid improves radiographic contrast by absorbing scattered radiation before it gets to the film.

Noise on an image appears as mottle, and the image has a grainy appearance (quantum mottle). This occurs when few photons are used to create an image on the X-ray film, since the main controlling factor for the number of photons from the X-ray tube is mAs. Low mAs will result in more noise compared to high mAs techniques. Furthermore there are other factors that affect noise such as the kVp. Less noise is produced when higher kVp techniques are used for the same mA settings. The technologist must therefore select the best possible factors in order to produce optimum image quality.

1.3.6 Radiation Dose Considerations

The radiation dose in film-screen radiography is affected by several factors including exposure technique factors, filtration, collimation and field size, scattered radiation grids, image receptors (detectors), and source-to-image receptor distance (SID); however, only the first factor will be reviewed here. While the dose is directly proportional to the mAs, meaning that if the mAs is doubled, the dose is doubled, it (dose) is proportional to the square of the change of the kVp. Higher kVp techniques will result in less absorbed dose in the patient [7].

1.3.7 Limitations of Film-Screen Radiography

Film-based radiography has been the workhorse of radiology ever since the discovery of X-rays in 1895, and despite the successful use for over 100 years and its present use in many departments today, one of the major problems with the radiographic imaging process is poor image quality if the initial radiation exposure has not been accurately determined. For example, if the radiation exposure is too high, the film is overexposed, and the processed image appears too dark, thus, the radiologist cannot make a diagnosis from such an image. Alternatively, if the radiation exposure is too low, the processed image appears too light and not useful to the radiologist, as shown in Fig. 1.4. In both of these situations, the images lack the proper image density and contrast and would have to be repeated to provide an acceptable image quality needed to make a diagnosis. Additionally, the patient would be subjected to increased radiation exposures due to repeat exposures.

There are other problems associated with film-based radiography. For example:

1. As a radiation detector, film-screen cannot show differences in tissue contrast less than 10%. This means that film-based imaging is limited in its contrast resolution. For example, while the contrast resolution (mm at 0.5% difference) for film-screen radiography is 10, it is 20 for nuclear medicine, 10 for ultrasound, 4 for computed tomography, and 1 for magnetic resonance imaging [6]. The spatial resolution (line pairs/mm) for radiography, however, is the highest of all the other imaging modalities and can range of 5–15 line pairs/mm [6]. This is the main reason why radiography has been so popular through the years.
2. As a display medium, the optical range and contrast for film are fixed and limited. Film can only display once, the optical range and contrast determined by the exposure technique factors used to produce the image. In order to change the image display (optical range and contrast), another set of exposure technique factors would have to be used, thus increasing the dose to the patient by virtue of a repeat exposure.
3. As an archive medium, film is usually stored in envelopes and housed in a large room. It thus requires manual handling for archiving and retrieval by an individual.

These problems can be overcome by a digital radiography imaging system.

1.4 Major Components of a Digital Radiography Imaging System

The major technical components of a digital radiography system are illustrated in Fig. 1.5 and include the data acquisition, computer data processing, image display and post-processing, image storage, image and data communications, and image and information management. Each of these will now be described briefly.

1.4.1 Data Acquisition

Data acquisition refers to the collection of X-rays transmitted through the patient. It is the first step in the production of the image. For digital radiography, special electronic detectors (digital detectors) are used and replace the X-ray film cassette used in film-based radiography. These detectors are of several types that utilize technologies to convert X-rays to electrical signals (analog signals). For example, while one type of detector will first convert X-rays into light, followed immediately by the conversion of the light into electrical signals, another type of digital detector will avoid the light-electricity conversion process and convert X-rays directly into electrical signals. The analog signals must be converted into digital data for processing by a digital computer. The conversion of analog signals is a function of the analog-to-digital converter (ADC).

1.4.2 Computer Data Processing

The ADC sends digital data for processing by a digital computer. The computer uses special software to create or build up digital images using the binary number system. While humans use the decimal number system (which operates with

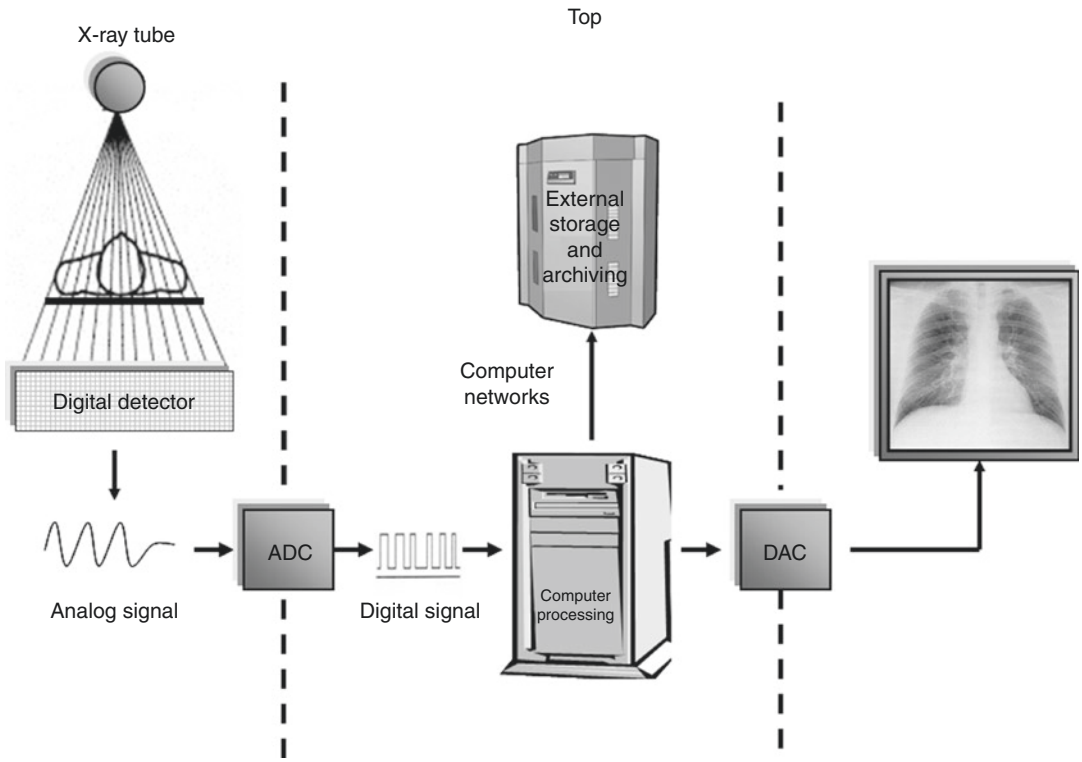

Fig. 1.5 The major technical components of a digital radiography system

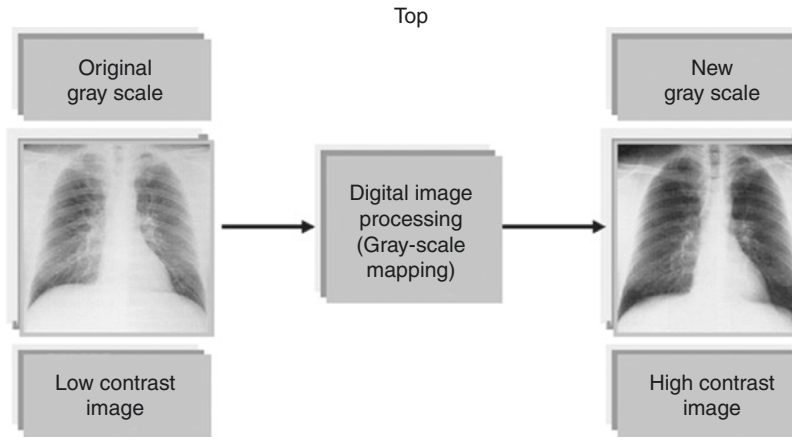

Fig. 1.6 The digital image processing tool of grayscale mapping can change the picture quality to suit the needs of the viewer

base 10, that is, 10 different numbers; 0,1,2,3,4,5,6,7,8,9), computers use the binary number system (which operates with base 2, that is, 0 or 1). These two digits are referred to as *bi*nary digi*ts* or *bits*. Bits are not continuous but rather, they are discrete units. Computers operate with binary numbers, 0 s and 1 s, discrete units that are processed and transformed into other discrete units. To process the word Euclid, it would have to be converted into digital data (binary representation). Thus the binary representation for the word Euclid is 01000101 01010101 01000011 01001100 01001001 01000100.

1.4.3 Image Display and Post-processing

The output of computer processing, that is, the output digital image must first be converted into an analog signal before it can be displayed on a monitor for viewing by the observer. Such conversion is the function of the digital-to-analog converted (DAC). The image displayed for initial viewing can be processed using a set of operations and techniques, referred to as post-processing techniques, to transform the input image into an output image that suits the needs of the observer (radiologist) in order to enhance diagnosis. For example, these operations can be used to reduce the image noise, enhance image sharpness, or simply change the image contrast or to stitch several images to form one image. The effect of one common and popular digital image processing tool, referred to as grayscale mapping, can be seen in Fig. 1.6.

1.4.4 Image Storage

The vast amount of images generated for the wide range of digital radiology examinations must be stored for not only retrospective analysis but also for medicolegal purposes. Today, various kinds of storage devices and systems are used for this purpose, such as magnetic tapes, disks, and laser optical disks, for long-term storage. In a PACS environment, for example, a storage system such as a RAID (redundant array of independent disks) is not uncommon. It is important to note that those images that are stored in a short-term archival system are deleted after a period of time defined by the institution.

1.4.5 Image and Data Communications

Image and data communications are concerned with the use of computer communication networks to transmit images from the acquisition phase to the display/viewing and storage phase. If the image transmission is within the hospital (Intranet), local area networks (LANs) are used.

If, however, the images have to be sent outside the hospital (Internet) to remote locations, networks such as wide area networks (WANs) must be used.

Picture Archiving and Communication Systems (PACS) are being used for storing/archiving and communicating images in the digital radiology department. In addition, information systems, such as the *radiology information systems* (RIS) and the *hospital information systems* (HIS), are now being integrated with the PACS via computer networks, using *communications standards* such as DICOM (Digital Imaging and Communications in Medicine) and HL-7 (Health Level-7), for effective management of patient information.

An important element of image and data communications is that of image compression [8]. The purpose of image compression is to reduce storage space (and hence costs) and decrease the image transmission time. Two popular compression methods for use in digital radiology are lossless or reversible compression, and lossy or irreversible compression. While in the former, there is no loss if information when the image is decompressed, the latter will result in some loss of information. The effect of these two compression methods on visual image quality are illustrated in Fig. 1.7.

1.4.6 Image and Information Management

Image and information management refers to the use of PACS and information systems such as RIS and HIS to manage the vast number of images and text data produced in a digital radiology department, using databases and file management software. While the RIS and HIS handle essentially textual information, specifically dealing with business operations for the entire hospital, the PACS handle images generated by the various digital imaging modalities (to be described subsequently).

1.5 Integrating the Healthcare Enterprise

Another important aspect of digital image acquisition, PACS-RIS-HIS integration is that of *Integrating the Healthcare Enterprise* (IHE). IHE is a model for ensuring that the standards for communication such as DICOM and HL-7 work effectively to facilitate integration. The concept of IHE had its origins in 1998 when two major organizations, the Radiological Society of North America (RSNA) and the Healthcare Information and Management Systems Society

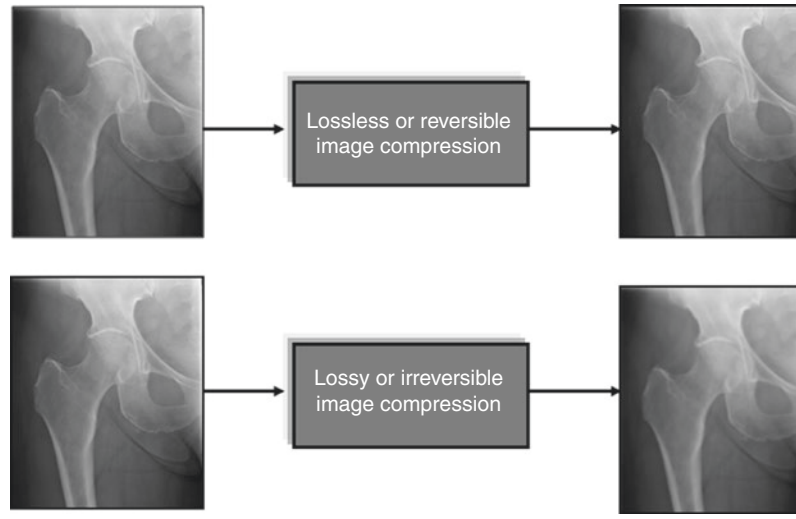

Fig. 1.7 The effects of two image compression methods on picture quality

(HIMSS), developed what they refer to as a technical framework based on three essential elements: a data model, an actor, and an integration profile.

1.6 Digital Radiography Modalities

Digital radiography includes several *imaging modalities* coupled to the PACS-RIS-HIS image and information systems and based on the technologies mentioned above. The imaging modalities include computed radiography (CR), flat-panel digital radiography (FPDR), digital mammography (DM), and digital fluoroscopy (DF). Imaging modalities and the PACS-RIS-HIS systems must be fully integrated for overall effective and efficient operations.

This section will provide an overall orientation by describing how these modalities work in the most fundamental way. Each of them will be described in detail in dedicated later chapters.

1.6.1 Computed Radiography

Computed radiography (CR) makes use of photostimulable or storage phosphors to produce digital images using existing X-ray imaging equipment. A digital computer is used to process data collected by radiographic means to produce digital images of the patient. In 1983, Fuji Medical Systems introduced a CR imaging system for use in clinical practice. Other companies such as Agfa, Kodak, Konica, and Cannon, to mention only a few, now manufacture CR imaging systems.

In CR, a photostimulable phosphor such as barium fluorohalide is coated on the plate referred to as an imaging plate (IP) that is housed in a cassette (similar in appearance to a film-screen cassette) to protect it from damage and exposure to foreign materials. The IP then is considered the digital detector in CR [6, 9].

The basic steps in the production of a CR image are shown in Fig. 1.8 and include the following:

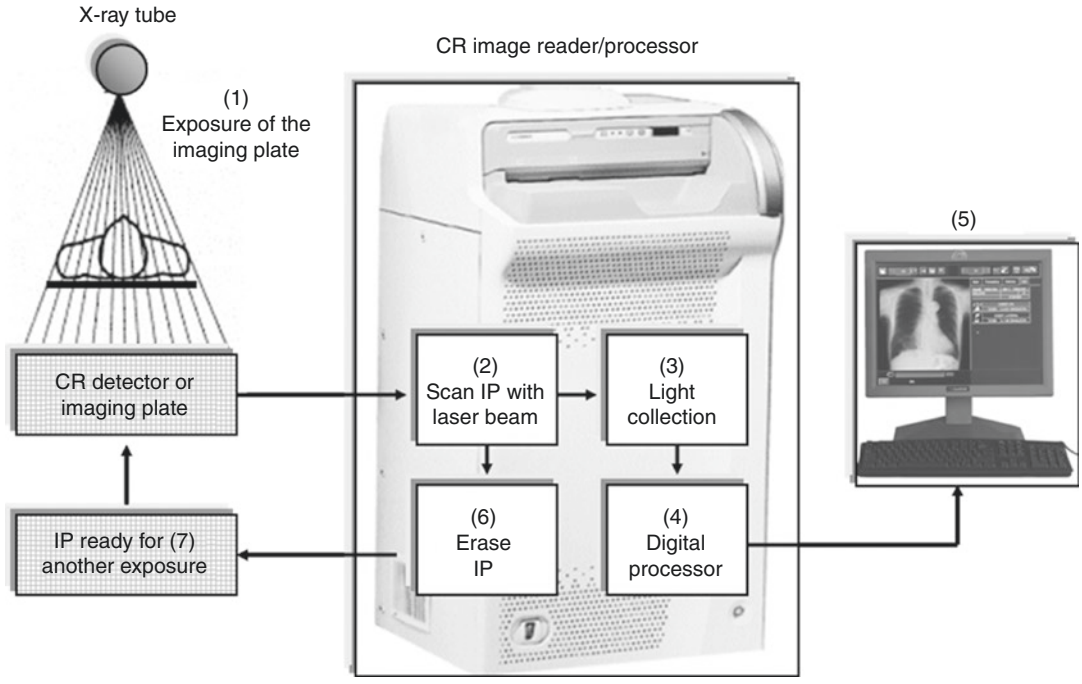

Fig. 1.8 The fundamental steps in the production of a CR image. See text for explanation of the numbers 1–7

1. The IP is exposed to X-rays, which causes electrons in the phosphor to move to another energy level, where they remain trapped to create a latent image.
2. The plate is then taken to the CR reader/processor (digital image processor) where it is scanned by a laser beam which causes the trapped electrons return to their original orbit, and in the process, light is emitted.
3. This light is collected by a light guide and sent to a photomultiplier tube (PMT). The output electrical signal from the PMT is subsequently converted into digital data.
4. A digital processor processes the digital data to produce a CR image.
5. The CR image is subsequently displayed for viewing.
6. The IP is exposed to a bright light to erase it (remove residual latent image).
7. The IP can now be used again.

One of the significant differences between CR and film-screen radiography is that exposure latitude of CR is about 10^4 times wider than that of the widest range of film-screen systems. This difference and others as well as similarities will be described further in the chapter on CR.

A major drawback of CR systems is that of their ability to image detail (spatial resolution). While the spatial resolution of a CR is about 3–5 line pairs/mm, it is about 10–15 line pairs/mm for film-screen radiographic imaging systems. The contrast resolution on the other hand is superior and can be manipulated for CR systems, while it is inferior and fixed for film-screen systems.

One important objective descriptor of digital image quality is the detective quantum efficiency (DQE). The DQE is a measure of how efficient a digital detector can convert the X-rays collected from the patient into a useful image [9]. The DQE for CR is much better than for film-screen systems. This means that CR can convert X-rays from the patient into useful data, over a wider exposure range compared with film-screen detectors. CR will be described in more detail in Chap. 3.

1.6.2 Flat-Panel Digital Radiography

Flat-panel digital radiography (FPDR) systems have been developed to overcome the shortcomings of CR systems. As the name implies, the digital detector is designed as a flat-panel, and it is totally different in design structure and function, compared to the CR detector (IP). Currently, there are two categories of flat-panel digital radiography imaging systems based on the type of detector used [6, 9], and they have been popularly referred to as:

1. *Indirect conversion* digital radiography systems.
2. *Direct conversion* digital radiography systems.

The basic components of these detectors and the steps in the production of an image for indirect and direct conversion systems are illustrated in Fig. 1.9a and b, respectively.

In Fig. 1.9a, X-rays are first converted into light using a phosphor such as cesium iodide. The emitted light from the phosphor falls upon a matrix array of electronic elements to create and store electrical charges in direct proportion to the X-ray exposure. The charges produce electrical signals, which are subsequently digitized and processed by a digital computer to produce an image.

Direct conversion digital radiography systems use detectors that convert X-rays directly to electronic signals. As shown in Fig. 1.9b, X-rays fall upon the photoconductor (e.g., selenium) which is coupled to a matrix array of electronic elements to produce electrical signals. These signals are digitized and processed by a digital computer to produce an image.

The second type of indirect conversion detector is shown in Fig. 1.10. This type uses an array of charge-coupled devices (CCDs) instead of an array of electronic elements as identified in the first type (Fig. 1.9a). The CCDs are coupled to a scintillator phosphor cesium iodide as shown. X-rays fall upon the phosphor to produce light, which then falls upon the CCD array that converts

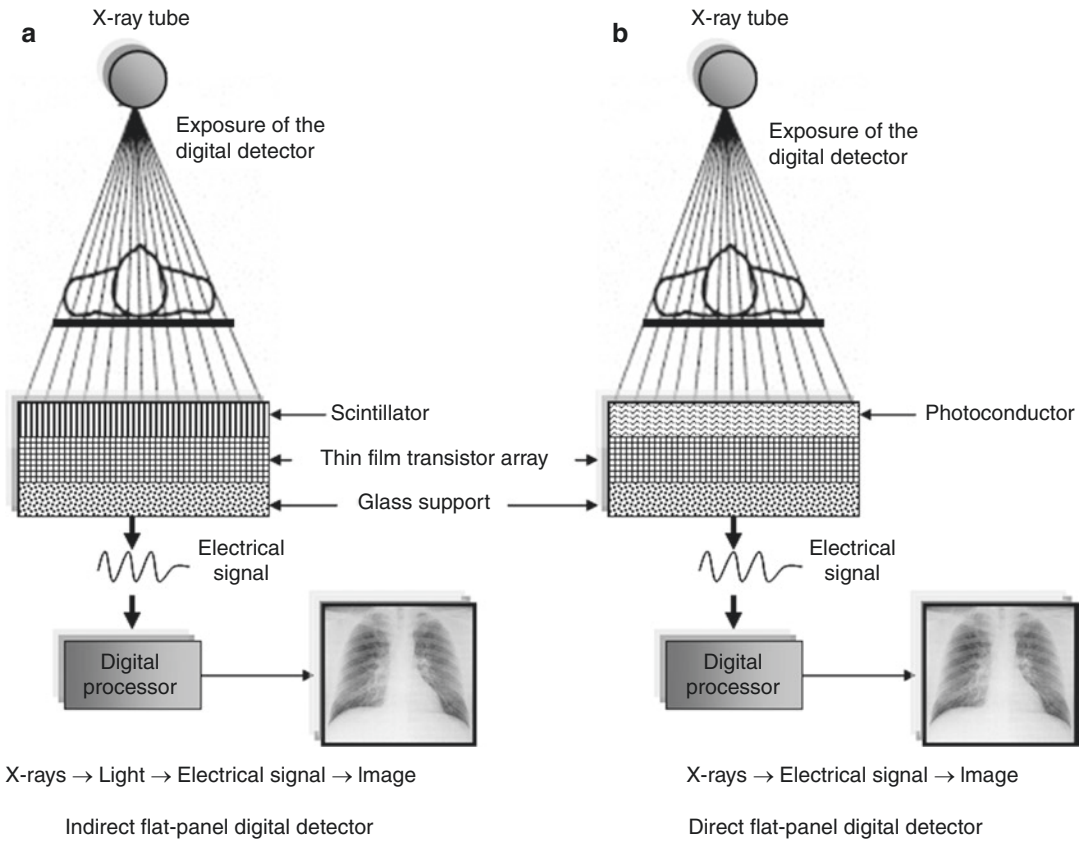

Fig. 1.9 The difference between two types of flat-panel digital radiography detectors, the indirect flat-panel system (**a**) and the direct flat-panel system (**b**)

the light into electrical signals, which are digitized and processed by a digital computer to produce an image.

It is important to note that after all electrical charges are read out; the flat-panel digital detector can be erased and ready to be used again. In addition, flat-panel detectors offer several advantages, including a high DQE and spatial resolution comparable to CR systems. Furthermore, the digital detectors used in CR and flat-panel digital radiography have a characteristic curve that is fundamentally different than the film characteristic curve. This curve is shown in Fig. 1.11. The digital detector output signal is linear with the input radiation exposure. This linear response produces wider exposure latitudes compared to the film characteristic curve, which has limited exposure latitudes (defined by their slopes). Herein lays a significant advantage of the digital detector. The wide exposure latitude of the digital detector will produce acceptable images even when the input exposure is low or high. As mentioned earlier, in film-screen radiography, while a low exposure will produce a light image, a high exposure will produce a dark image. These images would have to be repeated to produce acceptable film density, thus increasing the radiation dose to the patient. These will be discussed in more detail in the respective chapters on CR (Chap. 3) and flat-panel digital radiography (Chap. 5)

1.6.3 Digital Fluoroscopy

The application of digital image processing to fluoroscopy is referred to as *digital fluoroscopy*. One of the major goals of digital

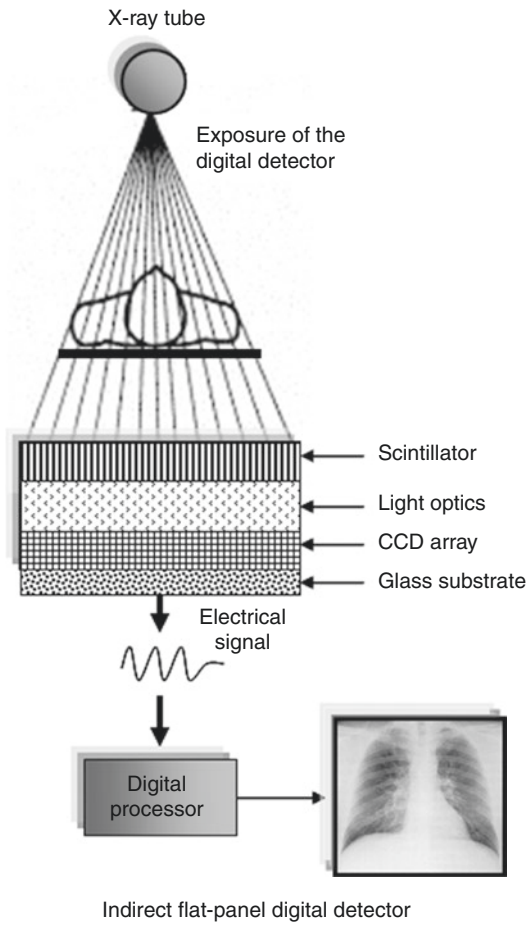

Fig. 1.10 An indirect flat-panel digital detector using a CCD array to produce digital images of a patient

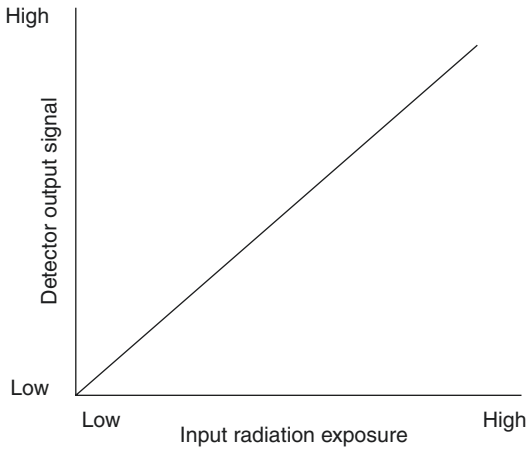

Fig. 1.11 The characteristic curve of a digital detector. The linear response provides a wider exposure latitude compared to a film-screen system

fluoroscopy is to improve the perception of contrast resolution compared to conventional fluoroscopy, by using digital image processing software. Other advantages of digital fluoroscopy are grayscale processing, temporal frame averaging, and edge enhancement, to mention only a few.

While radiography produces static images, fluoroscopy produces dynamic images acquired in real time to allow for the study of motion of organ systems and hollow internal structures such as the gastrointestinal tract, as well as blood circulatory system.

A conventional fluoroscopy imaging system is shown in Fig. 1.12 and consists of an fluoroscopic X-ray tube, an image intensifier tube (the radiation detector), associated optics, a television (TV) or CCD camera tube, and a television monitor for image display and viewing.

A digital fluoroscopy system on the other hand is shown in Fig. 1.13 and consists of all of the imaging components as for a conventional fluoroscopic imaging system, with a few differences. In digital fluoroscopy, the output signal (an electrical signal) from the TV/CCD video system is digitized by the ADC and sent to a digital computer for image processing. The detector in digital fluoroscopy is also the image intensifier tube, since it captures the radiation passing through the patient. Recently, flat-panel digital detectors are also being used in some digital fluoroscopy systems.

The application of digital fluoroscopy to angiography is referred to a digital subtraction angiography (DSA), whereby pre-contrast and post-contrast images can be digitally subtracted in real time. The goal of such operations is to improve the observer's perception of low-contrast vessels by subtracting or removing the tissues that interfere with visualization of vascular structures. Two such subtraction methods include temporal subtraction, in which images are subtracted in time, and energy subtraction, in which images are subtracted using different kilovoltages. Digital fluoroscopy will be described in detail in Chap. 7.

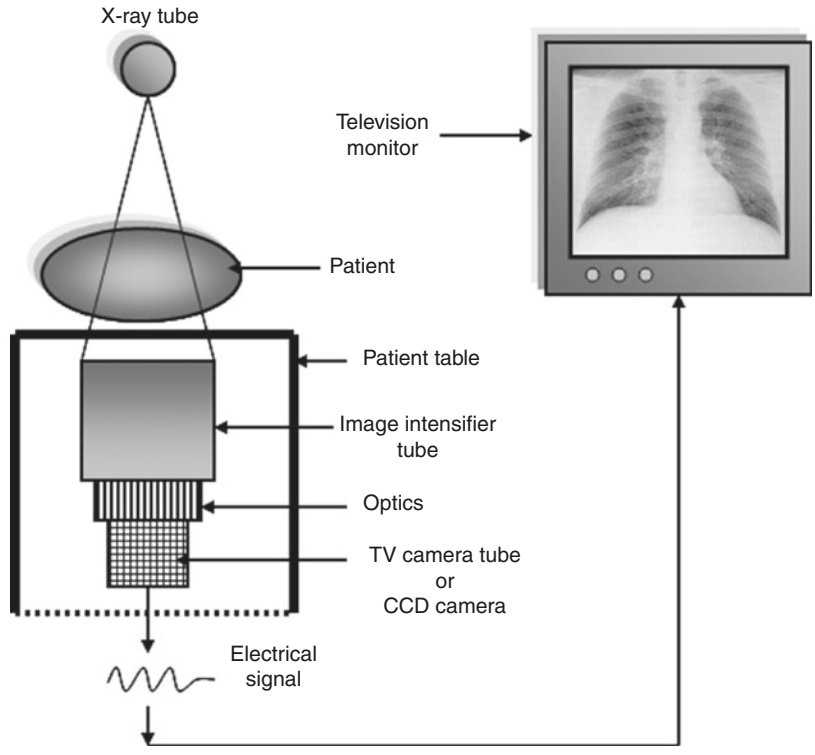

Fig. 1.12 The major technical components of a conventional fluoroscopic imaging system

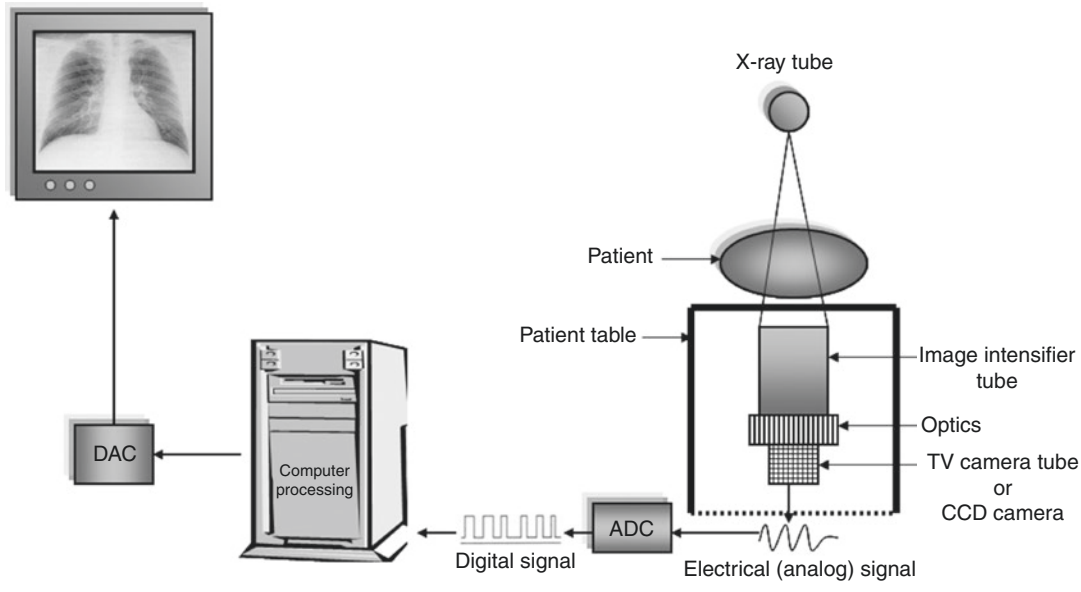

Fig. 1.13 The major technical components of a digital fluoroscopic imaging system. Note that a computer is used to process digital data from the analog-to-digital converter

1.6.4 Digital Mammography

Mammography is radiography of the breast and is often referred to as soft tissue imaging. *Film-screen mammography* requires a great deal of special technical considerations in order to detect breast cancer. Film-screen mammography suffers from all of the limitations of film-screen radiography described earlier. Digital mammography overcomes these limitations and offers several benefits as well. One such major benefit is that digital mammography allows the observer to use digital image processing tools to enhance diagnostic interpretation of the image.

Digital mammography systems currently utilize CR detectors and flat-panel digital detectors including direct and indirect conversion detectors, as well as CCD arrays to image the breast. Additionally, digital mammography uses computer-aided diagnosis (CAD) software to help radiologists enhance their detection of microcalcifications and malignant lesions and also provides the so-called "second-reader" approach [10, 11]. Digital mammography and its associated applications such as digital tomosynthesis will be described in Chaps. 8 and 9, respectively.

1.7 Picture Archiving and Communication Systems

The digital radiography modalities described above all produce digital images that must be displayed for interpretation, stored and archived for medicolegal purposes and retrospective analysis, and transmitted to remote locations to accommodate the needs of other users, such as surgeons in the operating room and emergency physicians within an institution or outside the institution. To perform this task efficiently and effectively requires the use of a Picture Archiving and Communication Systems (PACS). PACS have therefore become commonplace in digital radiology departments.

The backbone of a digital imaging department is a Picture Archiving and Communication System (PACS), a computer system for the management of the vast amount of digital images produced by the various digital image acquisition modalities. The term image management and communication systems (IMACS) has also been used to describe the various functions performed by this technology; however, the more common and popular term used today is PACS.

1.7.1 Definition of PACS

There are several definitions of PACS in the literature; however, one that is simple, and meaningful, and will be used in this book is:

> *….a comprehensive computer system that is responsible for the electronic storage and distribution of medical images in the medical enterprise. The system is highly integrated with digital acquisition and display devices and is often related closely to other medical information systems, such as the Radiology Information System (RIS) or Hospital Information System (HIS)* [12]

1.7.2 Major System Components

The major components of PACS are shown in Fig. 1.14 and include image acquisition devices, a PACS computer, devices called interfaces, and display workstations, all of which are connected and linked to the HIS and RIS, through digital communication networks.

The PACS computer is the heart of the system and is a "high-end" computer or server. Images and patient data (e.g., demographics) are then sent from the digital image acquisition modalities and the HIS and the RIS to the PACS computer, which has a database server as well as an archive system. Common forms of archiving in a PACS environment include laser optical disks, a redundant array of independent disks (RAID), and magnetic tapes.

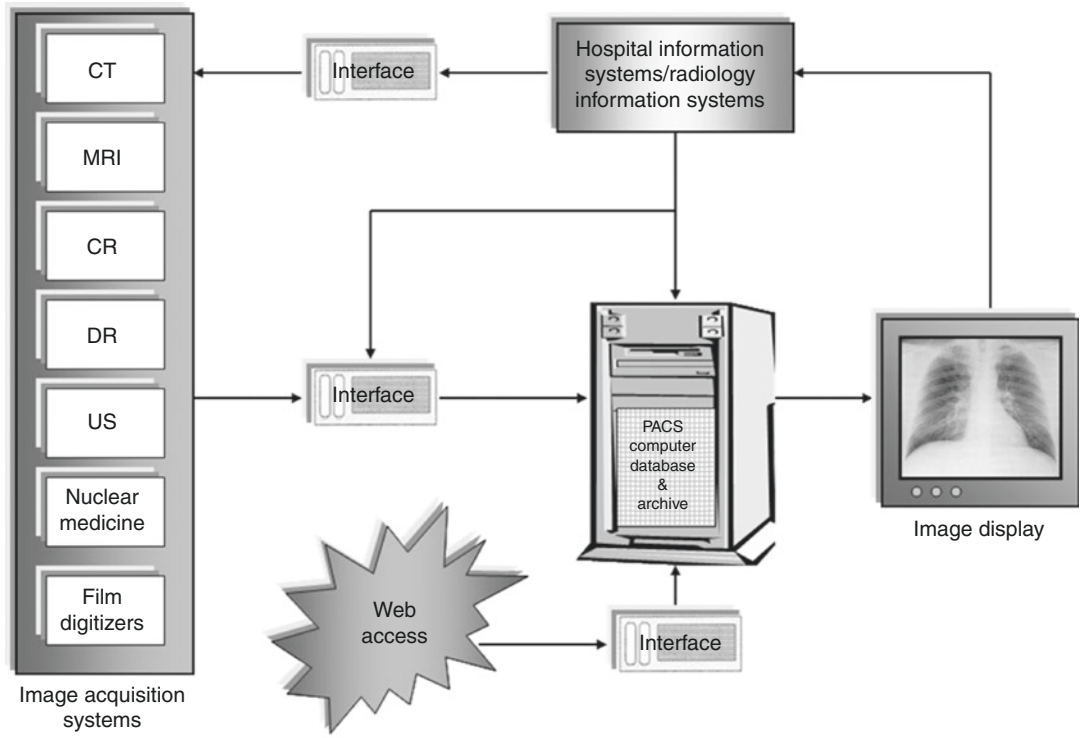

Fig. 1.14 A typical PACS system showing the major technical components required from acquisition of the image to the display of the image for viewing and interpretation. In this system users can access images stored in the PACS using web technology

Display workstations of soft-copy workstations, as they are often referred to, serve to display images on a monitor for the purpose of image interpretation. In addition, these workstations are of various types depending on the degree of use by the observer. While radiologists must have high-resolution diagnostic workstations that allow them to interpret images and provide a diagnosis of the patient's medical condition, technologists often use review workstations for general assessment of image quality before the images are sent to the PACS. A major feature of workstations is that they allow users to perform digital post-processing of images for the purpose of enhancing diagnosis. For example, images can be compressed to reduce storage space and decrease image transmission times.

In order for the PACS to work efficiently and effectively, other requirements must be met. One such mandatory requirement is the use of industry healthcare standards for data and image communications. Two standards that are currently used in a PACS environment are the DICOM (Digital Imaging and Communication in Medicine) and HL-7 (Health Level-7) standards. While DICOM is concerned primarily with images from the digital image acquisition modalities, HL-7 is concerned mainly with textual information from the HIS and RIS.

Since the PACS contain confidential patient data and information, it is essential that they be secured; hence, data security is of central importance in a digital hospital as well as in a PACS environment.

The interfaces shown in Fig. 1.14 serve to facilitate easy communications between the image acquisition modalities and the HIS/RIS, with the PACS computer, and also allow individuals to use the World Wide Web to access the

PACS computer as well. PACS will be described in more detail in Chap. 10.

1.8 Quality Assurance in Digital Radiography

Quality assurance (QA) and *quality control* (QC) procedures are effective strategies to ensure continuous quality improvement of a product. In radiology, QA/QC policies and procedures and related activities are all intended to:

1. Ensure that patients are exposed to minimum radiation using the ALARA (as low as reasonably achievable) philosophy.
2. Produce optimum image quality for diagnosis.
3. Reduce the costs of radiology operations.

For digital radiography systems, QA and QC continue to evolve and already special tools are now available to test the performance of digital imaging equipment, as well as the components of PACS. For example, QC tests for CR systems, for dark noise, exposure index calibration, and so on, are now becoming commonplace, and already technologists are performing these tests on a routine basis (AAPM, 2006).

The fundamentals of QC from digital radiography imaging systems will be described in Chap. 10.

1.9 Imaging Informatics

Digital imaging departments and the digital hospital now operate in the information technology (IT) domain. Digital image acquisition technologies, digital image processing, digital image display, storage, and archiving, as well as digital image communications utilize IT concepts. IT is a growing field, and it is becoming increasingly popular in all aspects of society. It involves both the use of computer technology and communications technology to solve problems in society including medical imaging and healthcare.

1.9.1 What Is Imaging Informatics?

Imaging informatics is defined by the Society for Imaging Informatics in Medicine (SIIM) as "... the study and application of processes of information and communications technology for the acquisition, manipulation, analysis, and distribution of image data [13]."

Medical imaging plays a significant role in healthcare since images can be used for diagnosis, assessment and planning (assesses the extent of a tumor and help to prepare an approach to management), guidance of procedures, communication, education, training, and research. These functions require the use of information technology (IT). IT is a growing field and is becoming increasingly popular in all aspects of society. IT involves the use of computer technology coupled with communications technology to solve problems in society, including medical imaging and healthcare. IT is now a major component of digital medical imaging.

IT involves such topics as information systems, standards for communicating both text and image data, computer communication networks, web technology, image and text handling, privacy, security, and confidentiality issues, and last but not least, digital image processing. More recently, the implementation of Picture Archiving and Communication Systems (PACS) in radiology departments is now commonplace. A more detailed description of Imaging Informatics will be presented in Chap. 11.

1.9.2 The Next Era in Imaging Informatics

The next era in medical imaging informatics includes at least six major topics that have received increasing attention in the literature. These include enterprise imaging, Big Data, cloud computing, machine learning and deep learning, and artificial intelligence. Each of these will be briefly stated here and will be elaborated on later in Chap. 12.

Imaging personnel and students must be aware of these concepts.

1.9.2.1 Enterprise Imaging

While medical imaging and hospitals are still using the conventional vendor-neutral archives (VNAs) and PACS, there has been increasing attention to move toward an enterprise imaging systems, with the goal of centralizing data and workload all in one system. This strategy will serve to improve efficiency in patient management and at the same time reduce operating costs to healthcare facilities. These efforts have led to the introduction and development of *enterprise imaging*, a term defined by the Healthcare Information and Management Systems Society (HIMSS) and the Society for Imaging Informatics in Medicine (SIIM) as "a set of strategies, initiatives and workflows implemented across a healthcare enterprise to consistently and optimally capture, index, manage, store, distribute, view, exchange, and analyze all clinical imaging and multimedia content to enhance the electronic health record" [14].

1.9.2.2 Cloud Computing

Another concept that has emerged in the domain of imaging informatics is that of *cloud computing*. This term refers to "the access of computing resources through the Internet for purposes of data storage, aggregation, synthesis, and retrieval, together with the capacity to act on the data with computational algorithms and software packages" [15]. Presently there are three cloud computing models: Infrastructure as a Service (IaaS), Platform as a Service (PaaS), and Software as a Service (SaaS). These will be described briefly in Chap. 12.

1.9.2.3 Big Data

The most recent "emerging science" in medical imaging informatics is the topic of "Big Data." "*Big Data* refers to extremely complex data sets characterized by the four Vs: *Volume*, which refers to the sheer number of data elements within these extremely large data sets; *Variety*, which describes the aggregation of data from multiple sources; *Velocity,* which refers to the high speed at which data is generated; and *Veracity*, which describes the inherent uncertainty in some data elements" [16]. Furthermore Big Data will have an influence on the practice of medicine including medical imaging such as "to enable personalized image interpretation, discovery of new imaging markers, value quantification, and workflow characterization" [16]. Potential applications will allow imaging departments to "venture confidently into the next era of informatics and data science" [16].

1.9.2.4 Machine Learning and Deep Learning

Machine learning is the study of computer algorithms which can learn complex relationships or patterns from empirical data and make accurate decisions [17]. It is an interdisciplinary field that has close relationships with artificial intelligence, pattern recognition, data mining, statistics, probability theory, optimization, statistical physics, and theoretical computer science. Applications of machine learning in medical imaging include medical image segmentation and registration, computer-aided detection and diagnosis systems for computed tomography (CT), and magnetic resonance imaging (MRI).

Deep learning on the other hand is yet another area that will be used more and more in medical imaging. Deep learning is a machine algorithm that "has the benefit that it does not require image feature identification and calculation as a first step; rather, features are identified as part of the learning process" [17].

1.9.2.5 Artificial Intelligence

Artificial intelligence (AI) is a component of computer science that "aims to mimic human cognitive functions…and can be applied to various types of healthcare data (structured and unstructured)…..and popular AI techniques include machine learning methods for structured data, such as the classical support vector machine and neural network, and the modern deep learning, as well as natural language processing for unstructured data" [18]. Major disease areas that use AI tools include cancer, neurology, and cardiology.

1.9.3 The Technologist as Informaticist

The total digital medical imaging department will require that technologists have additional skills related to IT. Already these departments have what is referred to as a PACS Administrator (a technologist in some departments/or an IT person in others) whose function is solely dedicated to ensuring the integrity of the PACS. To be effective and efficient in this role, technologists must educate themselves in not only aspects of IT but continue to learn more about the digital world of radiology, including digital image processing. This will lead to a new role function for the technologist as informaticist.

References

1. Mendelson DS, Rubin DL. Imaging informatics: essential tools for the delivery of imaging services. Acad Radiol. 2013;20:1195–212.
2. ABII. About imaging informatics. 2016. https://www.abii.org/About-Imaging-Informatics.aspx. Accessed Oct 2017.
3. American Association of Physicists in Medicine (AAPM) Report No 116. An exposure indicator for digital radiography. College Park, MD: American Association of Physicists in Medicine; 2009.
4. Gonzalez RC, Woods RE. Digital image processing. 4th ed. Toronto, Ontario: Pearson; 2018.
5. Seeram E. Digital radiography-an introduction. Clifton Park, NY: Delmar Cengage Learning; 2011.
6. Bushong S. Radiologic science for technologists. 11th ed. St Louis, MO: Elsevier; 2018.
7. Seeram E, Brennan P. Radiation protection in diagnostic imaging. Burlington, MA: Jones and Bartlett Learning; 2017.
8. Seeram E. Computed tomography: physical principles, clinical applications, and quality control. 4th ed. St Louis, MO: Elsevier; 2016.
9. Bushberg JT, Seibert JA, Leidholdt EM Jr, Boone JM. The essential physics of medical imaging. 3rd ed. Philadelphia, PA: Lippincott Williams & Wilkins; 2012.
10. Mahesh M. Digital mammography: an overview. Radiographics. 2004;24:1747–60.
11. Pisano ED, Yafee MJ. Digital mammography. Radiology. 2005;234:353–62.
12. Samei E, Seibert JA, Andriole K, Badano A, Crawford J, Reiner B, Flynn MJ, Chang P. General guidelines for purchasing and acceptance testing of PACS equipment. Radiographics. 2004;24:313–34.
13. SIIM. What is imaging informatics? 2017. https://siim.org. Accessed Oct 2017.
14. Roth C, Lannum L, Persons K. A foundation for enterprise imaging: HIMSS-SIIM collaborative white paper. J Digit Imaging. 2016;29:530–8.
15. Mell P, Grence T. The NIST definition of cloud computing, Special Publication. Gaithersburg, MD: U.S. Dept. of Commerce, National Institute of Standards and Technology; 2011. p. 800–145.
16. Kansagra AP, Yu J-PJ, Chatterjee AR, Lenchik L, Chow DS, Prater AB, et al. Big data and the future of radiology informatics. Acad Radiol. 2016;23:30–42.
17. Erickson BJ, Korfiatis P, Akkus Z, Kline TL. Machine learning. Radiographics. 2017;37:505–15.
18. Jiang F, Jiang Y, Zhi H, Dong Y, Li H, Ma S, et al. Artificial intelligence in healthcare: past, present and future. Stroke Vasc Neurol. 2017;2(4):230–43.

Digital Image Processing Concepts

2

Abstract

This chapter outlines the essential concepts of digital image processing which is now prevalent in digital radiography departments. A definition of digital image processing is first introduced followed by a brief historical development leading to its use in medical imaging. Image formation and representation are reviewed through a description of the nature of analog and digital images, as well as image domains. Furthermore, the five classes of image processing operations are identified, and characteristics of the digital image are described. These include matrix, pixels, voxels, bit depth, and image appearance. The three steps in digitizing an image, scanning, sampling, and quantization, are briefly outlined. The chapter concludes with a brief overview of four image processing operations, point, local, global, and geometric operations, and offers a point of view on digital image processing as an essential tool for digital radiography technologists.

2.1 Introduction

In Chap. 1, digital radiography was defined as projection radiography whereby a digital computer is used to process data acquired from the patient to generate digital images. These images are subsequently displayed for viewing and interpretation by an observer. While the image-viewing task of the technologist is to assess the overall image quality (image density, contrast, detail, noise, distortion, artifacts, and accurate positioning of the anatomy, etc.), the image-viewing task of the radiologist is mainly lesion detection. This task is only made possible based on the quality of the image submitted by the technologist. In addition, these images can be digitally manipulated to suit the viewing needs of the observer, through the use of digital image processing techniques, which have become commonplace in medical imaging. Sections of this chapter have been previously published from my PhD thesis, entitled, *Optimization of the Exposure Indicator as a Dose Management Strategy in Computed Radiography.* PhD Dissertation (Charles Sturt University, New South Wales, Australia, 2014).

The purpose of this chapter is to outline the essential elements of digital image processing as it relates to digital radiography. First, a definition of digital image processing will be given followed by a brief description of history and application areas of digital image processing. Secondly, image representation and the basic concepts of digital image processing will be discussed.

E. Seeram, *Digital Radiography*, https://doi.org/10.1007/978-981-13-3244-9_2

2.2 Definition of Digital Image Processing

Digital image processing simply means the processing of images using a digital computer. The data collected from the patient during imaging is first converted into digital data (numerical representation of the patient) for input into a digital computer (input image), and the result of computer processing is a digital image (output image).

These numerical images can be changed in several ways, to suit the viewing needs of the radiologist, in an effort to improve and enhance diagnostic interpretation and management of the vast amount of images acquired from patients.

Digital image processing is commonplace in the digital radiology department and is now part of the routine skills of technologists and radiologists.

2.3 Brief History

2.3.1 NASA

The history of digital image processing dates back several decades to the United States National Aeronautics and Space Administration (NASA) space program. The Jet Propulsion Laboratory, California Institute of Technology in Pasadena, also made notable developments to this field. NASA used digital image processing to process images beamed back to Earth from the Ranger spacecraft, to improve visualization of the surface of the moon, for example. Later benefits from the space program's research were applied to other fields such as photography, biology, forensics, defense, remote sensing, and medicine, including medical imaging.

2.3.2 Medical Imaging

All digital radiography imaging modalities, including computed radiography (CR), flat-panel digital radiography (FPDR), digital mammography (DM), and digital fluoroscopy (DF), utilize digital image processing as a central feature of their operations [1]. Additionally, other digital imaging modalities such as computed tomography (CT), magnetic resonance imaging (MRI), diagnostic ultrasound, and nuclear medicine also incorporate digital image processing as an essential tool to manipulate and enhance digital images to meet the viewing needs of all observers [2]. For this reason, it is important technologists and radiologists alike become well versed in the nature, scope, and principles of digital image processing. This chapter offers one small step in that direction.

2.4 Image Formation and Representation

One of the first steps in becoming aware and versatile with digital image processing is to understand the general nature of images. In this regard, Castleman [3], an image processing expert, uses set theory to classify images based on their form and method used to produce them. He conceptualizes images as a subset of all objects and that the image set contain subsets, such as visible images, non-visible images, optical images, and mathematical images. While visible images include photographs, drawings, and paintings, for example, non-visible images include temperature, pressure, and elevation maps. Optical images on the other hand include holograms, for example. Finally, mathematical images include continuous and discrete functions, as is clearly illustrated in Fig. 2.1.

Mathematical images are important in the world of digital imaging. For example, the classical sine wave is a continuous function that can be converted into a discrete function, both of which will generate two categories of images, namely, analog and digital images (Fig. 2.1).

2.4.1 Analog Images

If the image of the chest in Fig. 2.1 is scanned from left to right using a light source (positioned in front of the image) and a photomultiplier tube

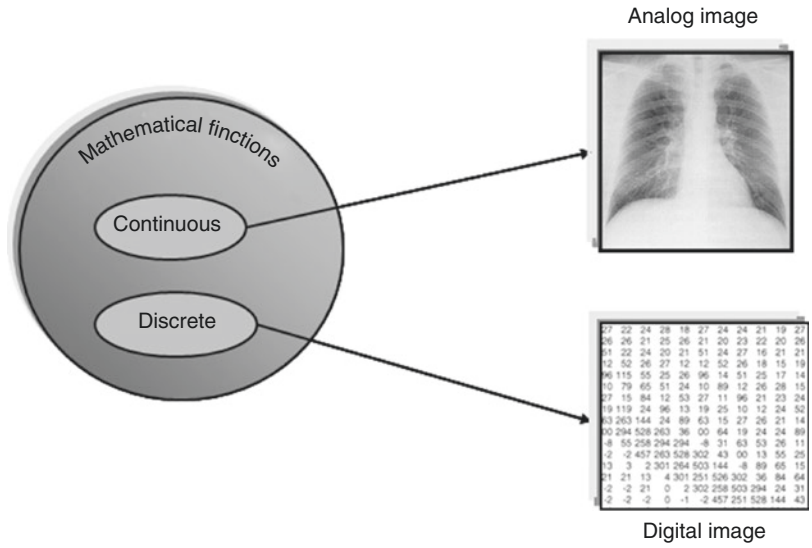

Fig. 2.1 Mathematical images include both continuous and discrete functions

(PMT-positioned behind the image), to detect the transmitted light, the light intensity will change continuously with respect to dark and bright spots on the image. The PMT would generate an output signal where its intensity varies continuously depending on the location of the light on the image. This signal is called an analog signal, and it represents the image scanned by the light source and the PMT. Such an image is called an analog image, because it is generated from a continuous function. It is important to note that in radiology, images displayed on monitors for viewing and interpretation are *analog images*.

2.4.2 Digital Images

In digital radiography, a *digital image* is a numerical representation of the patient, as is clearly illustrated in Fig. 2.2. For digital radiography, the following points are noteworthy:

1. The digital radiography detector outputs an analog signal (electrical signal).
2. This signal is sent to an *analog-to-digital converter* (ADC).
3. The ADC changes the continuous analog signal into discrete digital data. This is an important step in generating a digital image, simply because a digital computer requires discrete data (0s and 1s) for operation.
4. The result of computer processing is a digital image.
5. Since radiologists and technologists do not view numerical representations (digital images), these must be converted into a form suitable for human viewing. Hence, the digital image is converted into a visible physical image, an analog image.

In summary, digital image processing is defined by as "subjecting numerical representations of objects to a series of operations in order to obtain a desired result" [3]. This chapter will subsequently outline the elements of several operations.

2.4.3 Image Domains

The images obtained in radiology can be represented in two domains, based on how they are acquired. These include the spatial domain and the frequency domain [4, 5].

All images use a *right-handed X-Y coordinate system*, to identify the location of any number that makes up the image. In this case, the *X*-axis is used to describe the rows or lines placed on the image, while the *Y*-axis describes the columns.

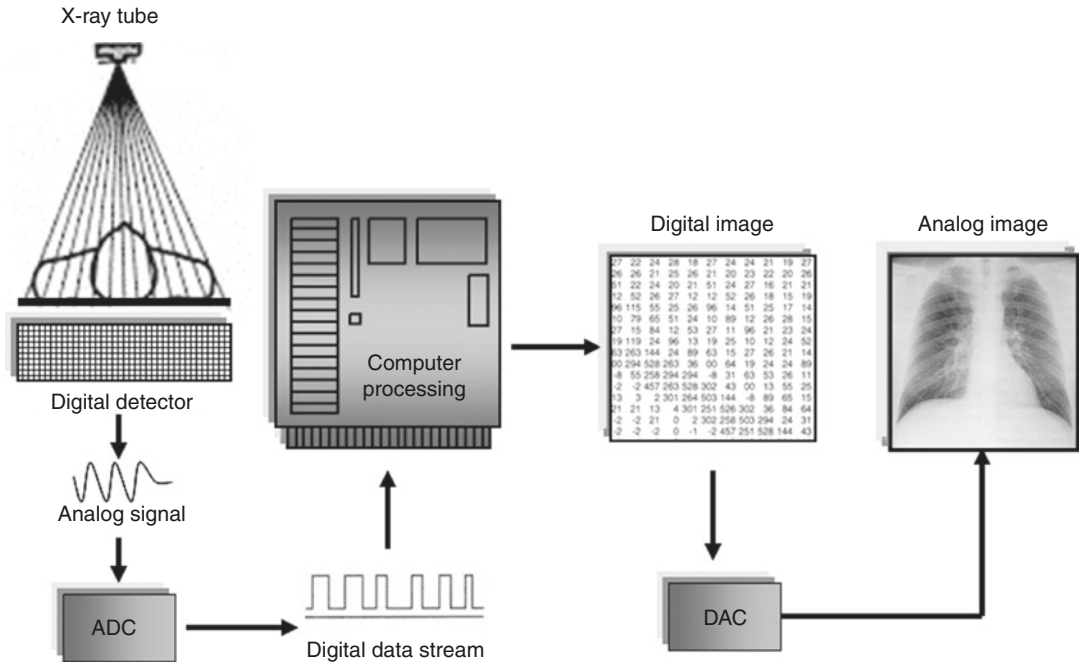

Fig. 2.2 A digital image is a numerical representation of the patient. The digital computer is used to process the data from the digital detector to produce a digital image

This is clearly shown in Fig. 2.3. For example, the first pixel in the upper left corner of the image is always identified as 0,0. The spatial location 5, 3 will describe a pixel that is located 5 pixels to the right of the left-hand side (L) of the image and 3 lines down from the top of the image. Such an image is said to be in the *spatial domain*.

Images can also be acquired in the *frequency domain* [6] as shown in Fig. 2.4 using the mathematical procedure of the Fourier transform (FT). The FT is not within the scope of this book and therefore will not be described here. The image is transferred back to the spatial domain using the inverse FT, to be viewed by observers such as radiologists and technologists.

One of the primary goals for doing this is to facilitate image processing that can enhance or suppress certain features of the image. For example, the image can be enhanced for sharpness in which case the low frequencies are suppressed, or it can be smoothed to enable better visualization of homogeneous structures by suppressing

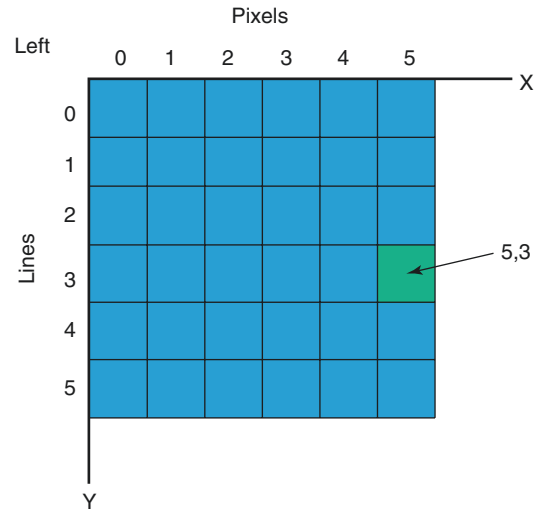

Fig. 2.3 A right-handed coordinate system for locating any pixel that makes up a digital image. In this case the gray box indicates a pixel located five pixels to the right of the left-hand side (L) of the image and three lines from the top of the image (From Seeram E.: Digital Image Processing. Radiologic Technology 2004, vol 75, No 6: 435–452) (Reproduced by permission of the ASRT)

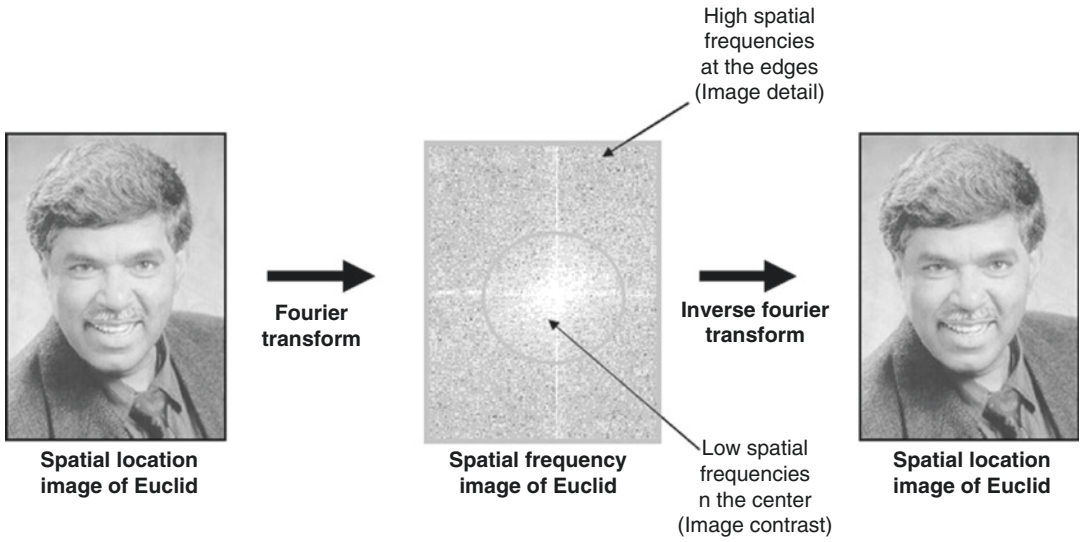

Fig. 2.4 An image of Euclid in the spatial domain (left) can be transformed using the Fourier transform into an image in the frequency domain (middle) and subsequently back into the spatial domain using the inverse Fourier transform

high frequencies via digital image processing in the frequency domain.

2.5 Classes of Digital Image Processing Operations

The operations used in digital image processing to transform an input image into an output image to suit the needs of the human observer are several. Baxes [7] identifies at least five fundamental classes of operations including image enhancement, image restoration, image analysis, image compression, and image synthesis. While it is not within the scope of this paper to describe all of these in any great detail, it is noteworthy to mention the purpose of each of them and state their particular operations. For a more complete and thorough description of these, the interested reader should refer to the seminal work of Baxes [7].

- *Image enhancement*: The purpose of this class of processing is to generate an image that is more pleasing to the observer. Certain characteristics such as contours and shapes can be enhanced to improve the overall quality of the image. The operations include contrast enhancement, edge enhancement, spatial and frequency filtering, image combining, and noise reduction.
- *Image restoration*: The purpose of image restoration is to improve the quality of images that have distortions or degradations. Image restoration is commonplace in spacecraft imagery. Images sent to Earth from various camera systems on spacecrafts suffer distortions/degradations that must be corrected for proper viewing. Blurred images, for example, can be filtered to make them sharper.
- *Image analysis*: Baxes [7] indicates that "the process of analyzing objects in an image begins with image segmentation operations, such as image enhancement or restoration operations. These operations are used to isolate and highlight the objects of interest. Then the features of the objects are extracted resulting in object outlines or other object measures. These measures describe and characterize the objects in the image. Finally, the object measures are used to classify the objects into specific categories." Segmentation operations are

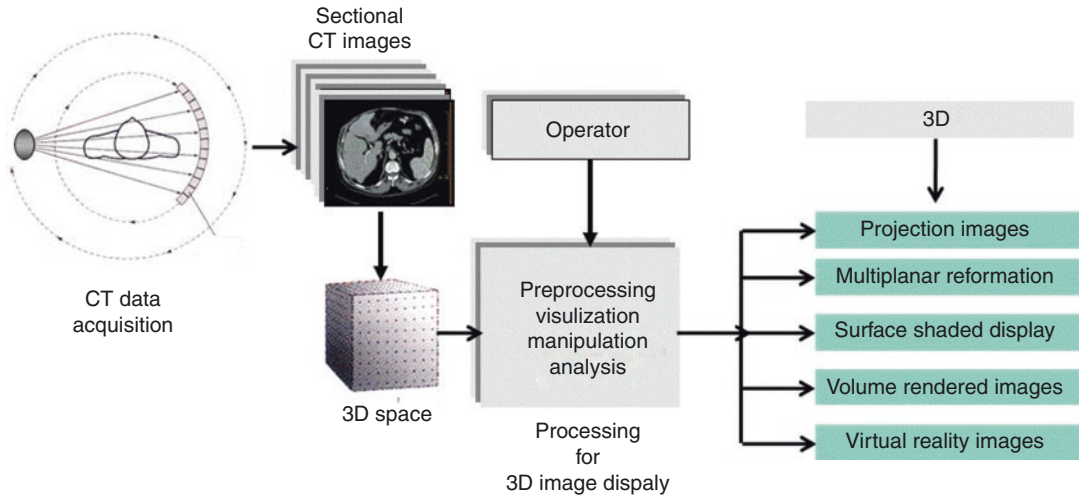

Fig. 2.5 3D imaging is an example of image synthesis. Note that new volumetric images (3D images) are created from a set of transverse sectional images collected from the imaging modality

used in three-dimensional (3D) medical imaging [2].

- *Image compression*: The purpose of image compression of digital images is to reduce the size of the image in order to decrease transmission time and reduce storage space. In general there are two forms of image compression, lossy and lossless compression. In lossless compression, there is no loss of any information in the image (detail is not compromised) when the image is decompressed. In lossy compression, there is some loss of image details when the image is decompressed. The latter has specific uses especially in situations when it is not necessary to have exact details of the original image [2]. A more recent form of compression that has been receiving attention in digital diagnostic imaging is that of wavelet (special waveforms) compression. The main advantage of this form of compression is that there is no loss in both spatial and frequency information.
- *Image synthesis*: These processing operations "create images from other images or non-image data. These operations are used when a desired image is either physically impossible or impractical to acquire, or does not exist in a physical form at all" [7]. Examples of operations are image reconstruction techniques, which are the basis for the production of CT and MR images, and 3D visualization techniques, which are based on computer graphics technology [2]. This is illustrated in Fig. 2.5.

2.6 Characteristics of the Digital Image

A digital image can be described with respect to several characteristics or fundamental parameters, including the matrix, pixels, voxels, and the bit depth [4].

2.6.1 Matrix

Apart from being a numerical image, there are other elements of a digital image that are important to our understanding of digital image processing. A digital image is made up of a two-dimensional array of numbers, called a *matrix*. The matrix consists of columns (M) and rows (N) that define small square regions called *pic*ture *el*ement*s* or *pixels*. The dimension of the image can be described by M and N, and the size of the image is given by the relationship

Table 2.1 Typical matrix sizes use in digital medical imaging

Digital imaging modality	Matrix size and typical bit depth
Nuclear medicine	128 × 128 × 12
Magnetic resonance imaging	256 × 256 × 12
Computed tomography	512 × 512 × 12
Digital subtraction angiography	1024 × 1024 × 10
Computed radiography	2048 × 2048 × 12
Digital radiography (flat-panel imagers)	2048 × 2048 × 12
Digital mammography	4096 × 4096 × 12

$$M \times N \times k \text{ bits}$$

When M = N, the image is square. Generally, diagnostic digital images are rectangular in shape. When imaging a patient using a digital imaging modality, the operator selects the matrix size, sometimes referred to as the *field of view* (FOV). Typical matrix sizes are shown in Table 2.1. It is important to note that as images become larger, they require more processing time and more storage space. Additionally, larger images will take more time to be transmitted to remote locations. In this regard, image compression is needed to facilitate storage and transmission requirements.

2.6.2 Pixels

The pixels that make up the matrix are generally square. Each pixel contains a number (discrete value) that represents a brightness level. The numbers represent tissue characteristics being imaged. For example, while in radiography and CT, these numbers are related to the atomic number and mass density of the tissues, in MRI, they represent other characteristics of tissues such as proton density and relaxation times.

The *pixel size* can be calculated using the relationship:

$$\text{Pixel size} = \text{FOV} / \text{matrix size}$$

For digital imaging modalities, the larger the matrix size, the smaller the pixel size (for the same FOV) and the better the spatial resolution. The effect of the matrix size on picture clarity can be seen in Fig. 2.6.

2.6.3 Voxels

Pixels in a digital image represent the information contained in a volume of tissue in the patient. Such volume is referred to as a *voxel* (contraction for *vo*lume *el*ement). Tissue voxel information is converted into numerical values and expressed in the pixels, and these numbers are assigned brightness levels, as illustrated in Fig. 2.7.

2.6.4 Bit Depth

The *bit depth* is the number of bits per pixel. An image that is described as having a bit depth of say 8 will have 256 (2^8) shades of gray. While Fig. 2.8 clearly illustrates the effect of the bit depth on image appearance, Table 2.1 also provides the typical bit depth for diagnostic digital images.

2.6.5 Appearance of Digital Images

The characteristics of a digital image, that is, the matrix size, the pixel size, and the bit depth, can affect the appearance of the digital image, particularly its spatial resolution and its density resolution.

The matrix size has an effect on the detail *or spatial resolution* of the image. The larger the matrix size (for the same FOV), the smaller pixel size, hence the better the appearance of detail. This is clearly illustrated in Fig. 2.6. Additionally, as the FOV decreases, without a change in matrix size, the size of the pixel decreases as well (recall the relationship pixel size = FOV/matrix size) thus improving detail. The operator selects a larger matrix size when imaging larger body parts, such as a chest, in order to show small details in the anatomy.

The bit depth has an effect on the number of shades of gray, hence the *density resolution* of the image. This is clearly apparent in Fig. 2.8.

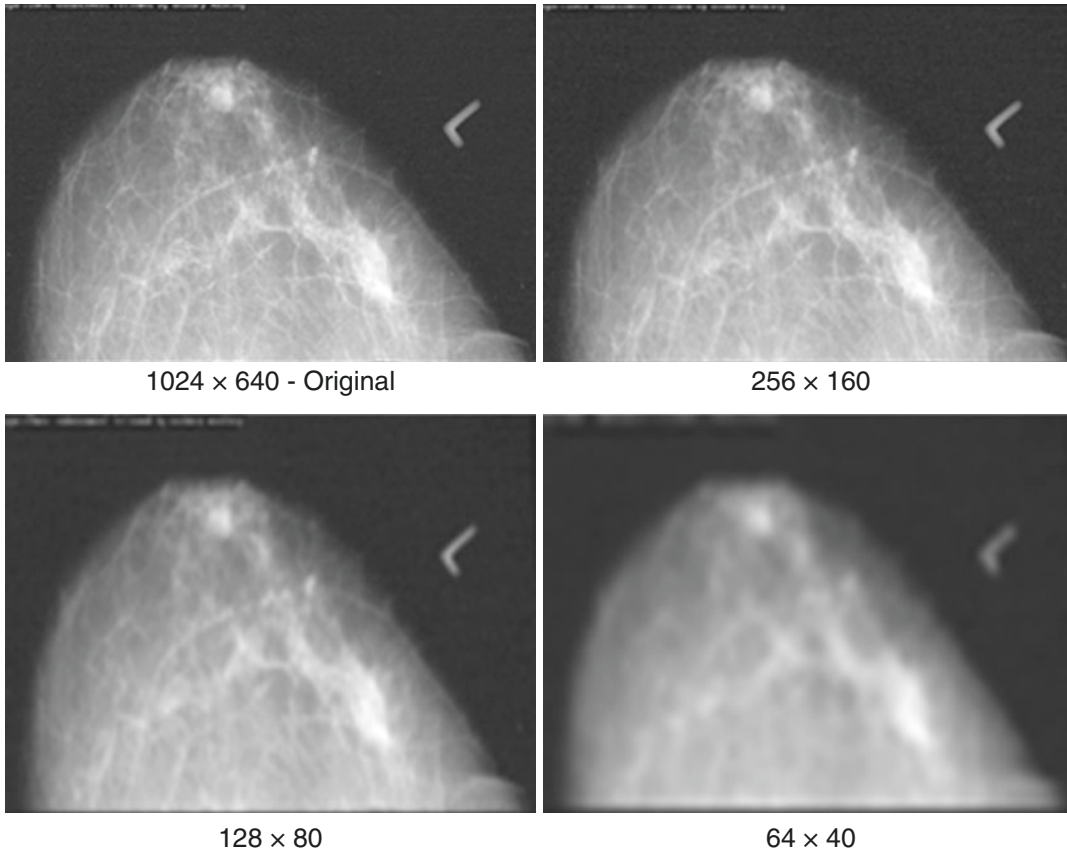

Fig. 2.6 The effect of matrix size on picture quality. As the matrix size increases for the same FOV, picture quality improves, that is the image becomes sharper (Images created by Bruno Jaggi, PEng-Biomedical Engineer-British Columbia Institute of Technology)

2.7 Steps in Digitizing an Image

In order to understand filmless imaging technology in the radiology department, it is first essential that we understand the fundamental steps to digitizing images, because similar steps apply to any digital imaging modality. There are three steps to digitizing an image: scanning, sampling, and quantization. Each of these will now be described briefly.

In *scanning*, the image is first divided into an array of small square regions called pixels. The second step, *sampling*, simply involves measuring the brightness level of each of the pixels using special devices such as a photomultiplier tube (PMT). The signal from the PMT is an analog signal (voltage waveform) that must be converted into a digital image for processing by a digital computer. The third step in digitizing an image is *quantization*. This is a process whereby the brightness levels obtained from sampling are assigned an integer (zero, a negative, or a positive number) called a gray level. The image is now made up of a range of gray levels. The total number of gray levels is called the gray scale.

The ADC (Fig. 2.2) plays an important role in the process of converting an analog signal into digital data for input into a digital computer. The

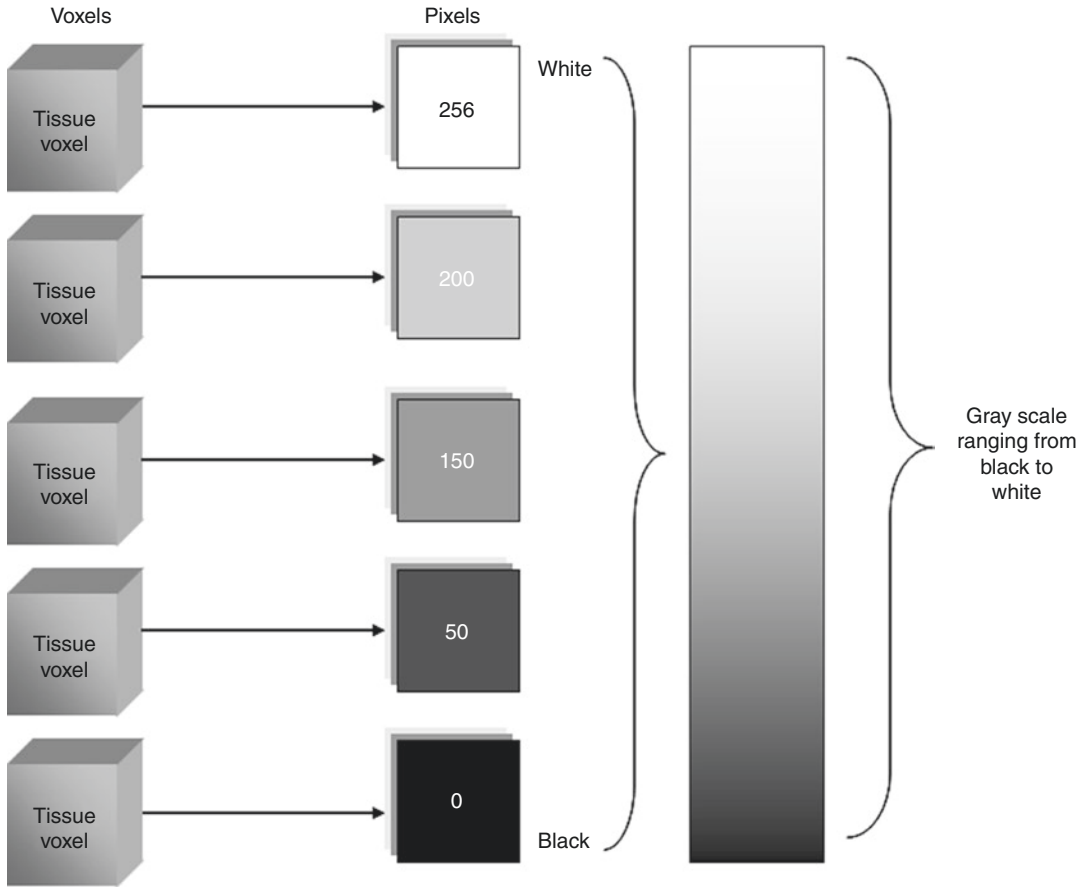

Fig. 2.7 The tissues contained in voxels are converted into numerical values that are represented in the pixels. These values are converted into a grayscale, where the lower numbers are represented as black and the higher numbers are represented as white

ADC consists of several components that will divide up the analog signal into equal parts. For example, s 2-bit ADC will convert the analog signal into 4 (2^2) equal parts, hence resulting in four gray levels. An 8-bit ADC will divide up the analog signal into 256 (2^8) parts, thus resulting in 256 gray levels. The image digitization steps are shown in Fig. 2.9.

Digital imaging modalities have 12- to 32-bit ADCs. The greater the bits, the more accurately the signals from the detectors can be digitized for a faithful reproduction of the original signal. This means that image quality is better with higher bit ADCs compared to lower bit ADCs.

2.8 Digital Image Processing Operations: General Concepts

Another important concept in digital image processing is processing itself. Both past and current-day processing technology includes a wide range of image processing algorithms for use in digital radiology [3–5, 7]. These include *point processing* operations such as *grayscale processing* (windowing, image subtraction, and temporal averaging), local processing operations (such as spatial filtering, edge enhancement, and smoothing), and global operations such as the

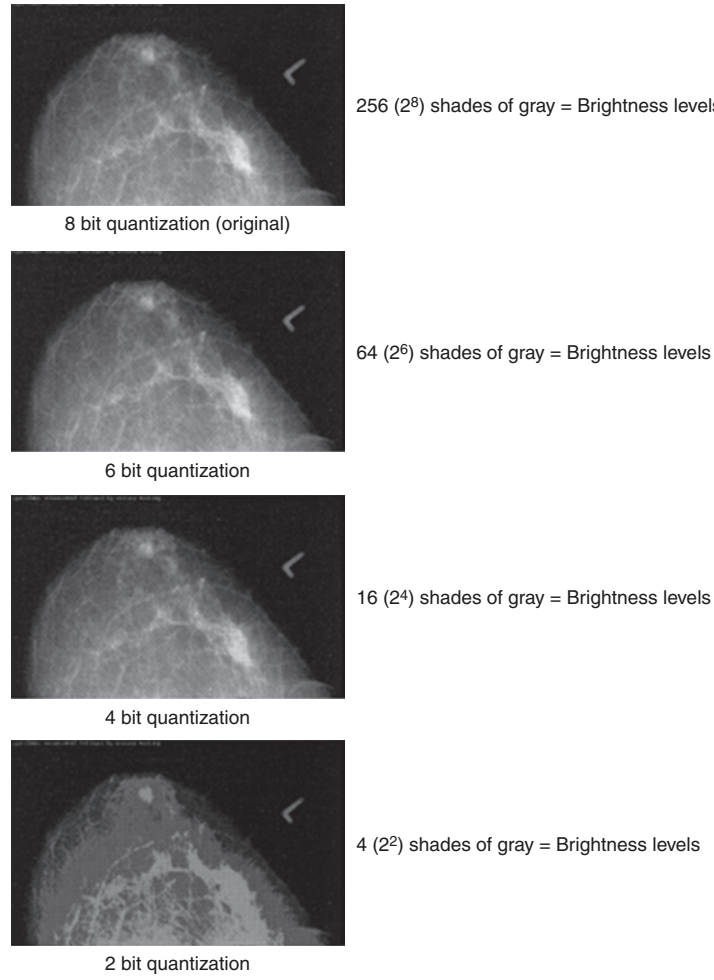

Fig. 2.8 The effect of bit depth on the shades of gray in an image (density resolution). The higher the bit depth, the greater the shades of gray (Images created by Bruno Jaggi, PEng-Biomedical Engineer-British Columbia Institute of Technology)

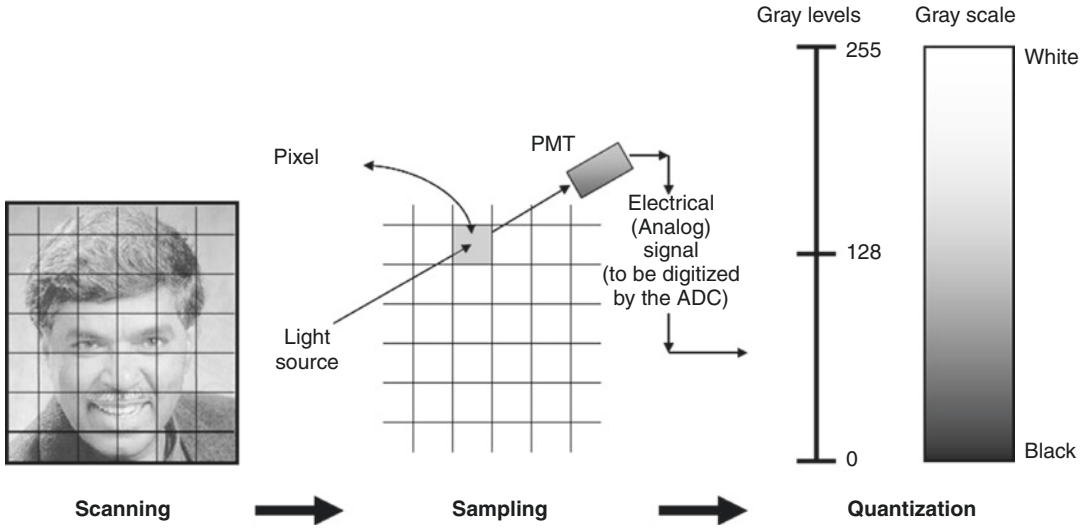

Fig. 2.9 The essential steps in digitizing an image. See text for further explanation (From Seeram E.: Digital Image Processing. Radiologic Technology 2004, vol 75, No 6: 435–452) (Reproduced by permission of the ASRT)

Fourier transform (FT). It is not within the scope of this article to describe the details of these processing algorithms; however, a conceptual overview of these operations for single images (as opposed to multiple images) will be presented since a number of them are used in digital radiology, several examples of which are shown in Table 2.2.

Table 2.2 Common digital image processing operations used in diagnostic digital imaging technologies (From Seeram E.: Digital Image Processing. Radiologic Technology 2004, vol 75, No 6: 435–452) (Reproduced by permission of the ASRT)

Digital imaging modality	Common image processing operations
Computed tomography	Image reformatting, windowing, region of interest (ROI), magnification, surface and volume rendering, profile, histogram, collage, image synthesis
Magnetic resonance imaging	Windowing, region of interest (ROI), magnification, surface and volume rendering, profile, histogram, collage, image synthesis
Digital subtracting angiography/ digital fluoroscopy	Analytic processing, subtraction of images out of a sequence, grayscale processing, temporal frame averaging, edge enhancement, pixel shifting
Computed radiography/ digital radiography	Partitioned pattern recognition, exposure field recognition, histogram analysis, normalization of raw image data, grayscale processing (windowing) spatial filtering, dynamic range control, energy subtraction, etc.

2.8.1 Point Processing Operations

A common and most often used *point processing operation* in medical imaging is one illustrated in Fig. 2.10, which shows that one pixel value in the input image is mapped onto the same pixel location in the output image and depends on the input pixel value. The entire image is then transformed using what is referred to as "pixel point process" [7]. Grayscale mapping (also referred to as "contrast stretching," "contrast enhancement," "histogram modification," "histogram stretching," or simply "windowing") is most often used in digital imaging to change the contrast and brightness of an image displayed for viewing on a display monitor. The term used by medical imaging professionals such as technologists and radiologists is *windowing*.

Image contrast and brightness transformations can be done using a variety of processing techniques, and it is noteworthy here to describe the fundamental concepts of at least two common methods used in digital radiology, the look-up table (LUT) method and the windowing method. Before describing each of these, the concept of a histogram must be understood.

2.8.1.1 Histogram

A *histogram* is a graph of the number of pixels in the entire image or part of the image having the same gray levels (density values), plotted as a function of the gray levels, as shown in Fig. 2.11. Changing the histogram of the image can alter

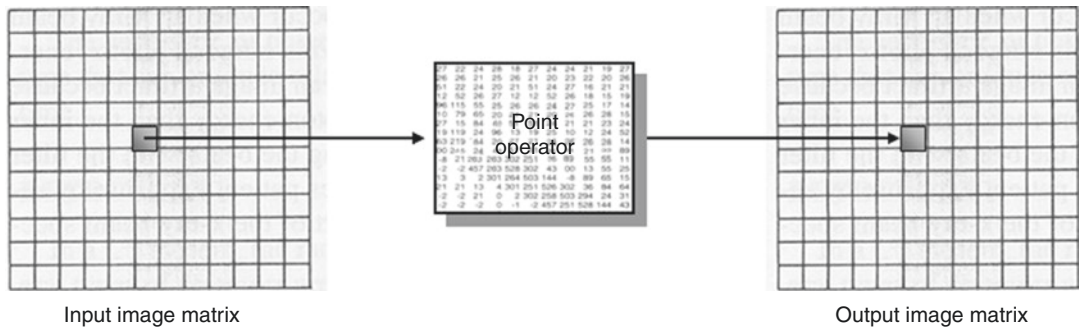

Fig. 2.10 The point processing digital operator. See text for further explanation

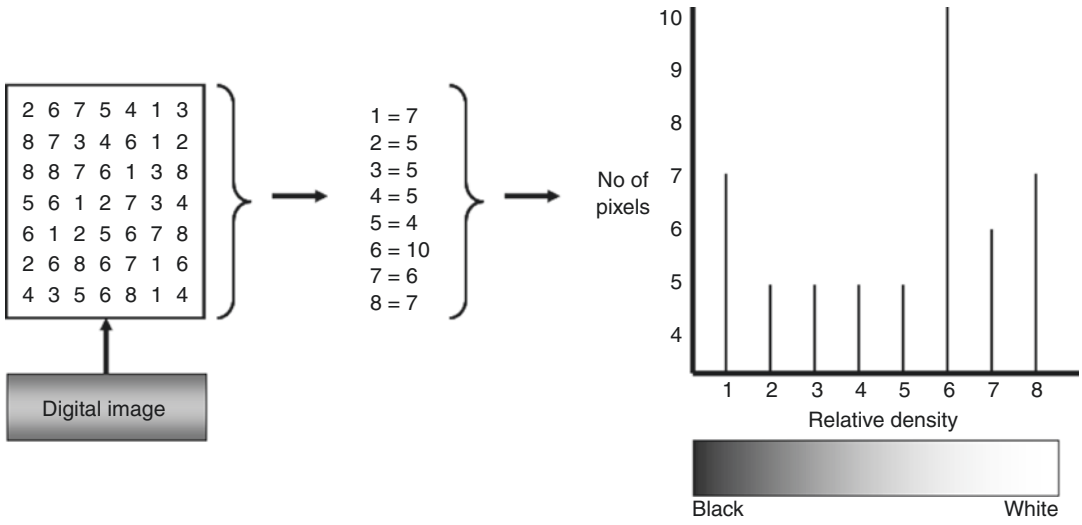

Fig. 2.11 A digital image can be converted into a histogram where the number of pixels is plotted as a function of the relative density. Adjusting the histogram will change the density of the image

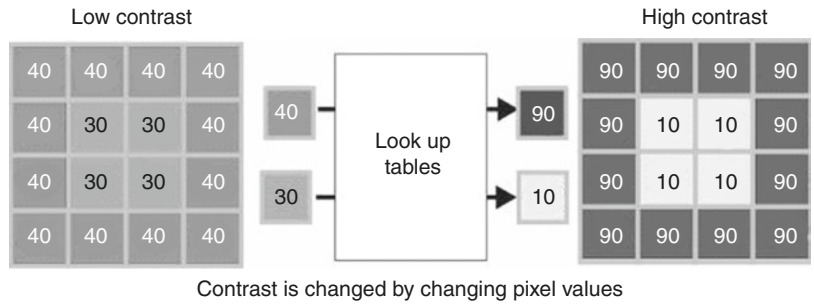

Fig. 2.12 Image manipulation allows the operator to change a low-contrast image into a high-contrast image using a look-up table (Courtesy of Dr Perry Sprawls, PhD)

the brightness and contrast of the image. If the histogram is modified or changed, the brightness and contrast of the image will change as well. This operation is called histogram modification or histogram stretching. While a wide histogram implies more contrast, a narrow histogram will show less contrast. On the other hand, if the values of the histogram are concentrated in the lower end of the range of values, the image appears dark, as opposed to a bright image, in which case the values are weighted toward the higher end of the range of values.

2.8.1.2 Look-Up Table

Figure 2.12 illustrates how a low-contrast image can be transformed into a high-contrast image using what is popularly referred to as a look-up table (LUT). Using this method, the LUT changes input pixel numerical values to corresponding output pixel values resulting in an image with different contrast and brightness (than the input image) (Figs. 2.13 and 2.14).

Digital radiographic imaging systems (e.g., such as CR) utilize a wide range of LUTs stored in the system, for the different types of clinical examinations (e.g., chest, spine, pelvis, extremities). The operator should therefore select the appropriate LUT to match the part being imaged. An important point to note here is the following: since digital radiographic detectors have wide exposure latitude and a linear response, the image displayed without processing may appear

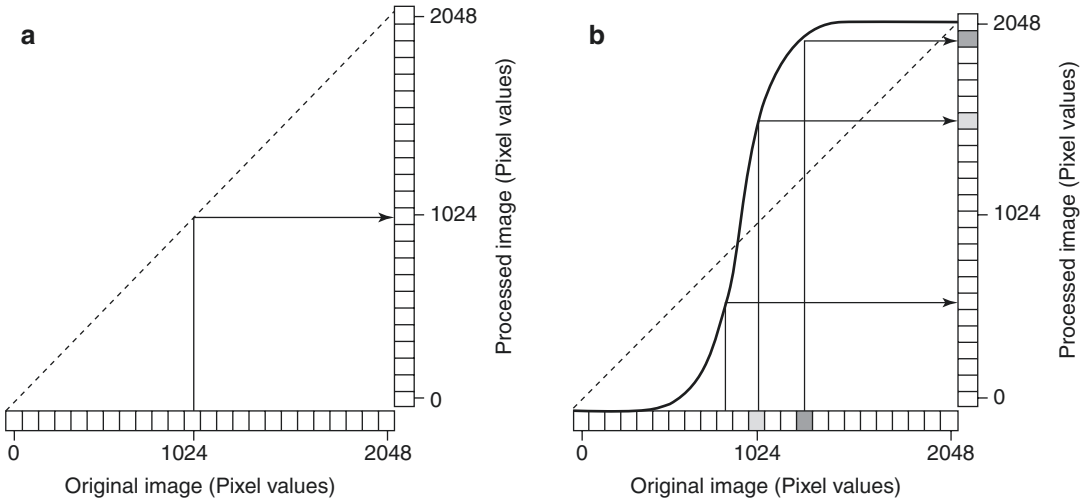

Fig. 2.13 The pixel values in the input image plotted as a function of the pixel values in the output image result in a straight line (**a**); however, a LUT curve is possible (**b**). The latter is the characteristic H & D curve for film (Courtesy of Dr Perry Sprawls, PhD)

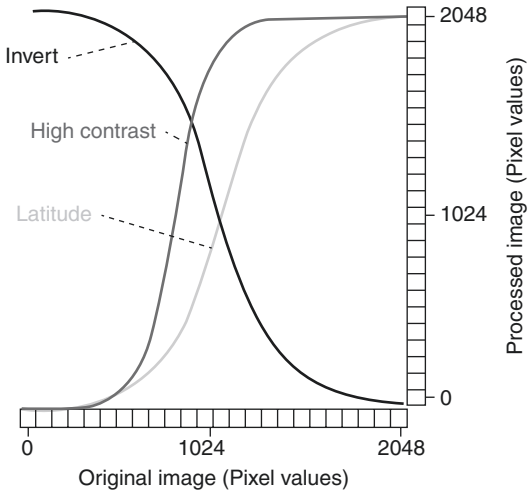

Fig. 2.14 Other LUT curves are possible when the input pixel values are plotted as a function of the output pixel values, such as the three curves shown (Courtesy of Dr Perry Sprawls, PhD)

as a low-contrast image. A processing example for a chest image using the LUT is shown in Fig. 2.15.

2.8.1.3 Windowing

The digital image processing technique known as *windowing* is also intended to change the contrast and brightness of an image. A digital image is made up of numbers, and by definition, the range of the numbers is the *window width* (WW), and the center of the range is defined as the *window level* (WL) [8]. The range of the pixel values (gray levels) and displayed image contrast range are shown in Fig. 2.16. The WW is used to change the contrast of the image, and the WL is used to change the image brightness. Whereas Fig. 2.17 illustrates the effect of a narrow and wide WW on image contrast, Fig. 2.18 shows the effect of WL changes on image brightness. Increasing WL values render the image darker since more of the lower numbers are displayed.

2.8.2 Local Processing Operations

A *local processing operation* is one in which the output image pixel value is obtained from a small area of pixels around the corresponding input pixel, as is illustrated in Fig. 2.19. Because a small area of pixels, or group of pixels is used, these operations are also referred to as area or group processes. A notable example is that of frequency filtering, as described earlier in this chapter. Frequency processing can sharpen, blur, smooth, and reduce the noise of an image [9].

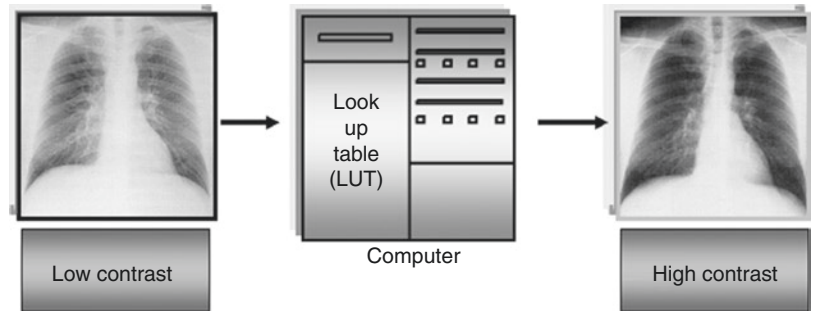

Fig. 2.15 The effect on a chest image of using a LUT to convert a low-contrast image into a high-contrast image

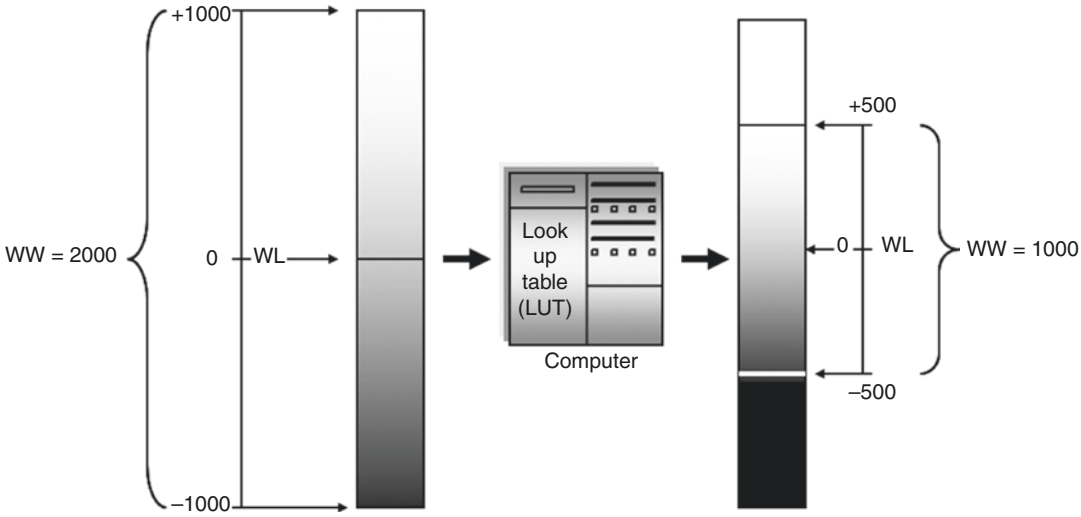

Fig. 2.16 A graphical illustration of the effect of using a LUT to change a wide range of pixel values (WW=2000/WL=0) into a narrow range of pixel values (WW=1000/WL=0)

2.8.2.1 Convolution

Figure 2.20 illustrates the technique of convolution (filtering in the spatial domain) where a group of input image pixels surrounding the specific input pixel (P5 in this diagram) is used to change the value of the output pixel value to arrive at a new value for P5 in the output image (using a weighted averaging). The group of pixels used to calculate this average is called the convolution kernel. A typical kernel size is a 3×3 matrix[2]. During processing, the kernel scans across the entire image, pixel by pixel.

Each input image pixel, the pixels surrounding it, and the kernel are used to calculate the corresponding output pixel value.

2.8.2.2 High-Pass Filtering

Figure 2.21 shows an example of the effect of using a high-pass filter. Such a filter is used to sharpen or for edge enhancement. The high-pass filter suppresses the low frequencies, and the result is a much sharper image than the original.

2.8.2.3 Low-Pass Filtering

Figure 2.22 shows how an image can be blurred effectively, using a low-pass filter. This is referred to as image smoothing. The output image appears blurred compared to the input image. In this type of filtering, the output image noise is reduced, and the image sharpness is compromised.

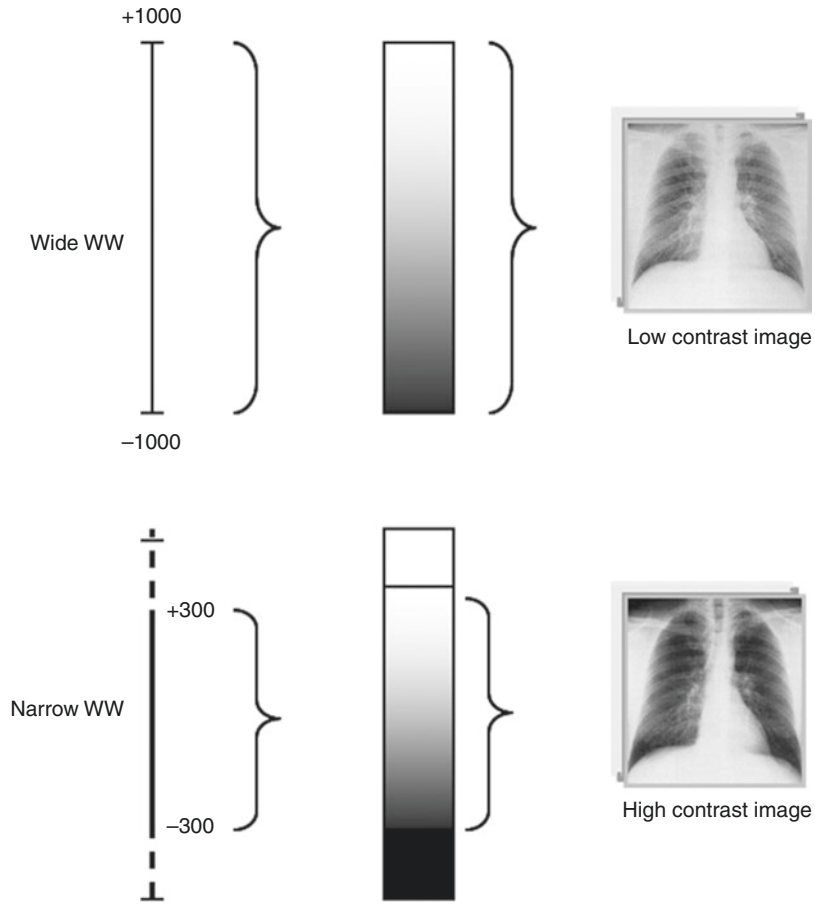

Fig. 2.17 The effect of the changing the window width (WW), on the contrast of a chest image. A narrow WW results in an image of higher contrast, while the window level (WL) remains fixed (From Seeram E.: Digital Image Processing. Radiologic Technology 2004, vol 75, No 6: 435–452) (Reproduced by permission of the ASRT)

2.8.2.4 Unsharp (Blurred) Masking

The effect of *unsharp* (blurred) *masking* on the image appearance is clearly shown in Fig. 2.23. First a low-pass filter is used to blur the image and subsequently subtracts this blurred image from the original image. The result is a sharper image than the original.

2.8.3 Global Processing Operations

The term "global" implies that all the pixels in the entire input image are used to change the value of a pixel in the output image. The conceptual framework for *global processing operations* is shown in Fig. 2.24. One popular global operation is to use the FT in filtering images in the frequency domain rather than in the spatial domain [4, 7]. These techniques can process images for edge enhancement, image sharpening, and image restoration. It is not within the scope of this chapter to describe the details of global processing operations.

2.8.4 Geometric Operations

Another class of image processing operations that are sometimes used in digital radiology is that of *geometric operations*. These techniques

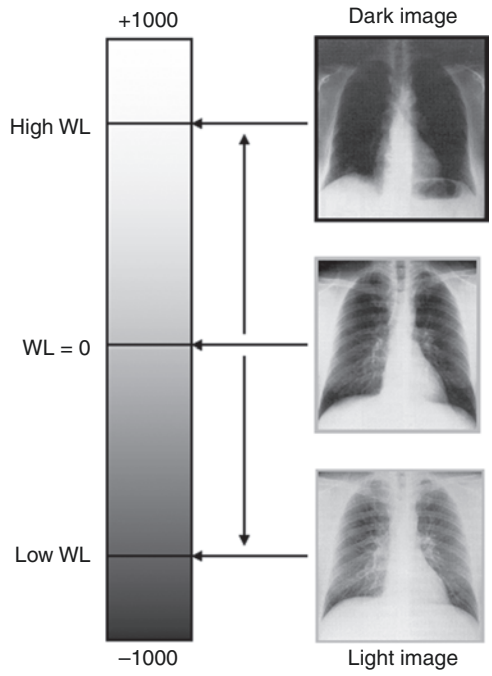

Fig. 2.18 The effect of changing the WL on image brightness. As the WL decreases (low WL), the image increases in brightness, for a fixed WW (From Seeram E.: Digital Image Processing. Radiologic Technology 2004, vol 75, No 6: 435–452) (Reproduced by permission of the ASRT)

allow the user to change the position or orientation of pixels in the image rather that the brightness of the pixels. Geometric operations result in the scaling, sizing, rotation, and translation of images, once again, to enhance diagnosis.

2.9 Digital Image Processing: An Essential Tool for Technologists

Digital image processing are techniques that allow the user to change the appearance of a digital image displayed on a monitor for viewing and interpretation. These techniques, for example, allow both the grayscale and the sharpness of the image to be manipulated to enhance diagnostic interpretation. Image post-processing is now commonly used by all medical imaging professionals who have the responsibility for assessing image quality and diagnostic interpretation of all images produced in an examination.

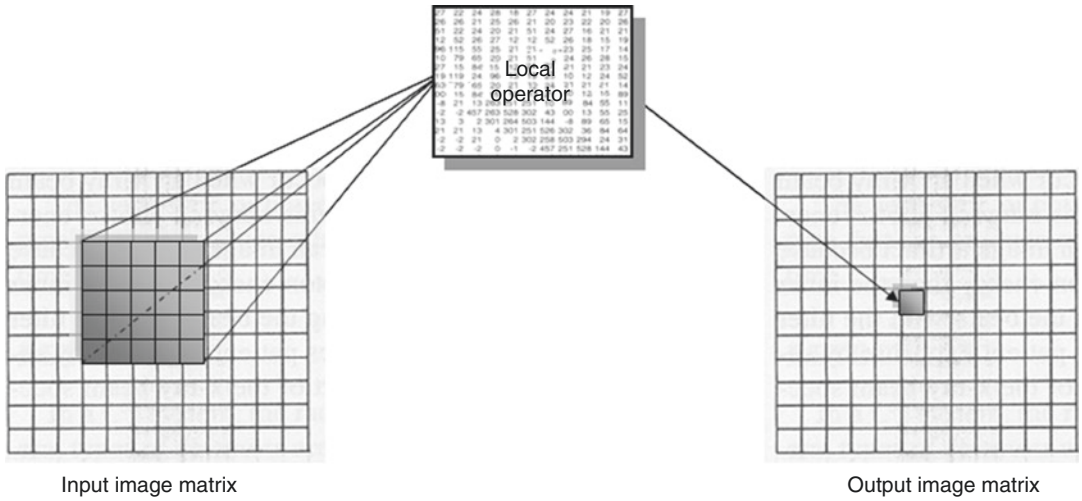

Fig. 2.19 A local processing digital operator uses a defined region of pixels in the input image to change one pixel value in the output image

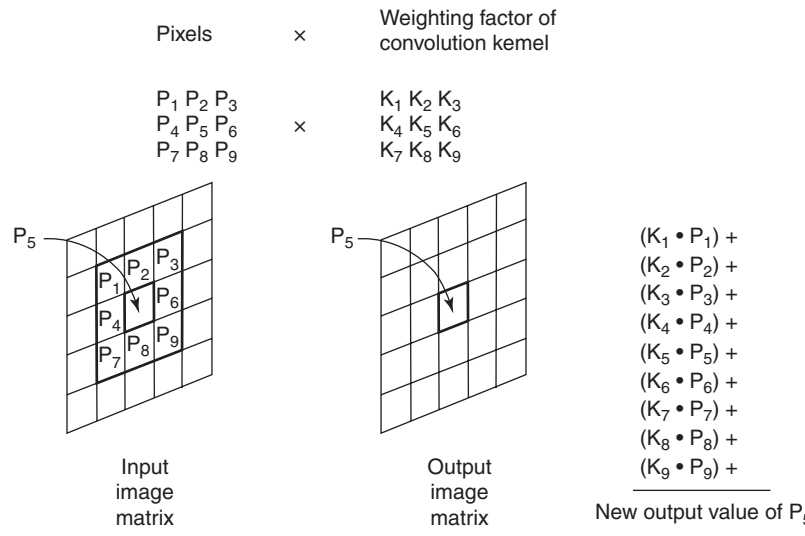

Fig. 2.20 An example of the convolution technique. See text for further explanation (From Seeram E.: Digital Image Processing. Radiologic Technology 2004, vol 75, No 6: 435–452) (Reproduced by permission of the ASRT)

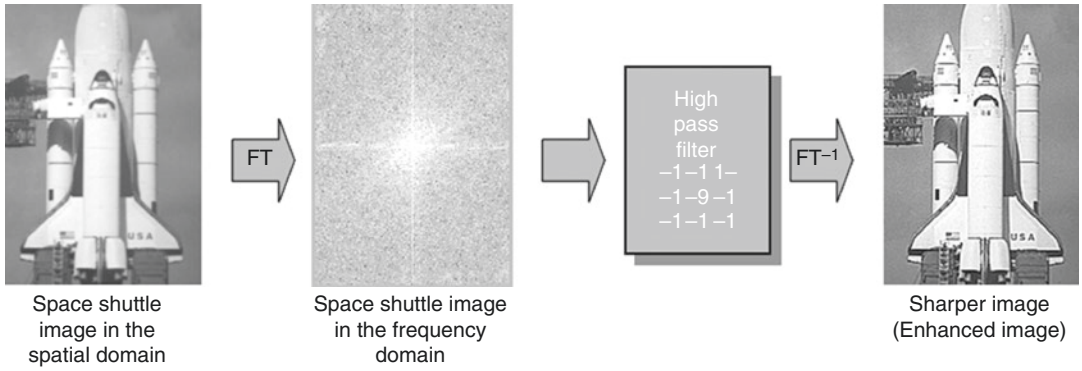

Fig. 2.21 The effect of using a high-pass digital filter on the quality of the output image. Note that in order to use this filter, the input image is first changed from a spatial location image into a spatial frequency image. The filter operates on the frequencies to sharpen the output image

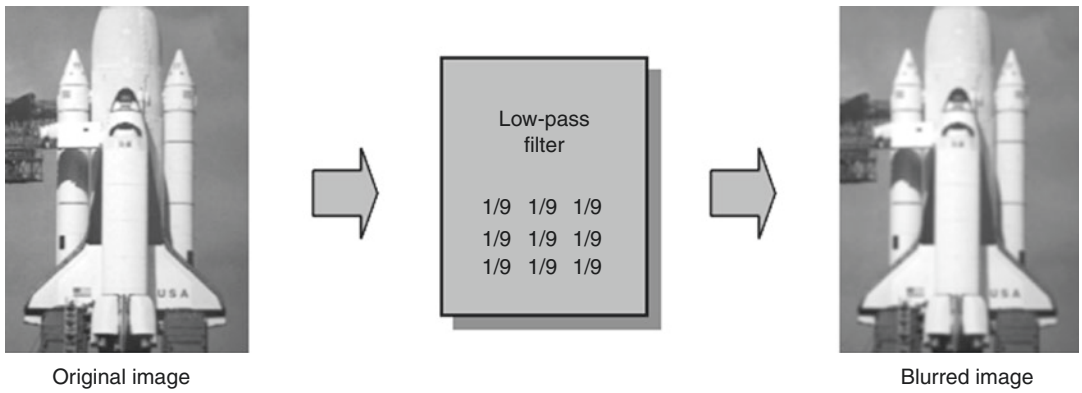

Fig. 2.22 The effect of a low-pass filter on the quality of the output image. In this case the output image is blurred

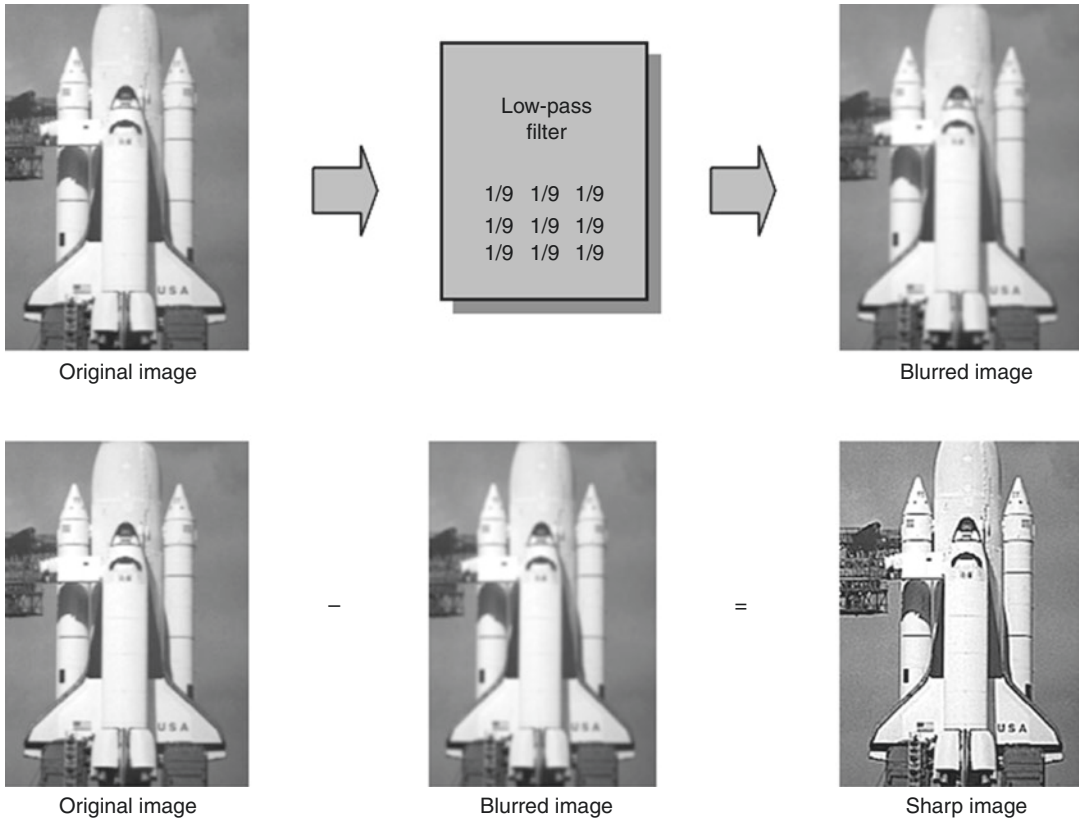

Fig. 2.23 The digital processing technique of unsharp masking can be used to sharpen the output image

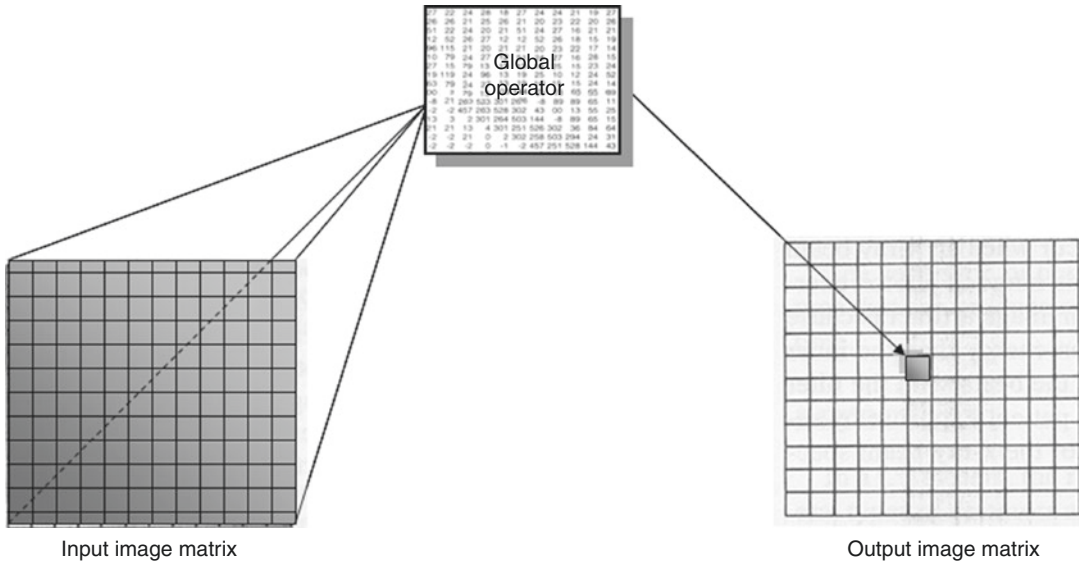

Fig. 2.24 A global processing digital operator uses all of the pixels in the input image to change one pixel value in the output image

References

1. Seeram E. Digital radiography-an introduction. Clifton Park, NY: Delmar Cengage Learning; 2011.
2. Seeram E. Computed tomography: physical principles, clinical applications, and quality control. St Louis, MO: Elsevier; 2016.
3. Castleman KR. Digital image processing. Englewood Cliffs, NJ: Prentice Hall; 1996.
4. Gonzalez RC, Woods RE. Digital image processing. 4th ed. Toronto, Ontario: Pearson; 2018.
5. Solomon C, Breckon T. Fundamentals of digital image processing. West Sussex, UK: Wiley-Blackwell; 2011.
6. Wolbarst AB, Capasso P, Wyant AR. Medical imaging: essentials for physicians. Hoboken, NJ: Wiley-Blackwell; 2013.
7. Baxes GA. Digital image processing: principles and applications. New York: John Wiley & Sons, Inc; 1994.
8. Bushong SC. Radiologic science, physics, biology and protection. 11th ed. St Louis, MO: Elsevier; 2017.
9. Jensen JR. Introductory digital image processing: a remote sensing perspective. 4th ed. Glenview, IL: Pearson Education Inc; 2015.

3 Computed Radiography: Physics and Technology

Abstract

This chapter describes the essential physics and technical considerations of computed radiography (CR). Firstly, a brief history of CR is reviewed including terms synonymous with CR, followed by a description of three main processes involved in CR including image acquisition, image processing, and image display, storage, and communication. Secondly, the basic physics of CR is described focusing on the nature of photostimulable storage phosphor (PSP), latent image formation and photostimulable luminescence (PSL), and PSL characteristics. Thirdly, the major topic outlined is the technological aspects of CR. These include the structure of CR imaging plate (IP), the IP imaging cycle, CR reader types and scanning technologies, features of the CR workstation, and computer networking and CR. Furthermore, image processing is elaborated and described in terms of preprocessing and post-processing operations. Fourthly, radiation exposure control in CR is explained in terms of the IP response to exposure, exposure indicators, and exposure indicator guidelines. Fifthly, an important topic described in this chapter deals with image quality descriptors including spatial resolution, density resolution, noise, and detective quantum efficiency. In addition, an overview of the nature of CR image artifacts is briefly reviewed. The final topic included in this chapter addresses continuous quality improvement (CQI) which includes quality assurance (QA) and quality control (QC).

3.1 Introduction

The basic principles of *computed radiography (CR)* were introduced in Chap. 1. In review, CR is a digital radiographic imaging modality, whereby a digital detector is used to capture X-rays transmitted through the patient. CR is based on the phenomenon of photostimulable luminescence which is exhibited by photostimulable phosphors. The CR digital detector is made of a photostimulable phosphor which, when struck by X-rays, creates a latent image. The latent image is rendered visible when the detector is scanned by a laser beam to produce light (photostimulable luminescence) that is subsequently converted into electrical signals. These signals are digitized and processed by a digital computer that produces the CR image, using special digital image processing algorithms.

The purpose of this chapter is to describe in detail the CR imaging system in terms of the basic physics of the CR detector and the technology needed to produce a CR image. In addition, the chapter will outline the essential elements of image processing, exposure control, image quality, and image artifacts and continuous quality improvement in CR.

E. Seeram, *Digital Radiography*, https://doi.org/10.1007/978-981-13-3244-9_3

3.2 Terms Synonymous with CR

In the medical imaging literature, several other terms have been used to refer to CR. These include photostimulable luminescence (PSL), storage phosphor radiography (SPR), digital luminescence radiography (DLR), photostimulable storage phosphor (PSP) radiography, and digital storage phosphor (DSP) radiography [1]. The term that has become commonplace, however, is computed radiography (CR), and, therefore, it will be used throughout this book.

3.3 A Brief History of CR

The history of CR is linked to *photostimulable phosphors* and the phenomenon of *photostimulable luminescence* and can be traced back to the 1600s with the discovery of the Bolognese stone (glowing stone) in Italy. Later, in the 1800s, Becquerel worked on the notion of de-excitation of atoms by optical means. This was followed by several notable developments, as shown in Fig. 3.1.

It was in 1983 that commercialization of CR imaging systems for use in diagnostic radiology began, when Fujifilm (Tokyo, Japan) introduced

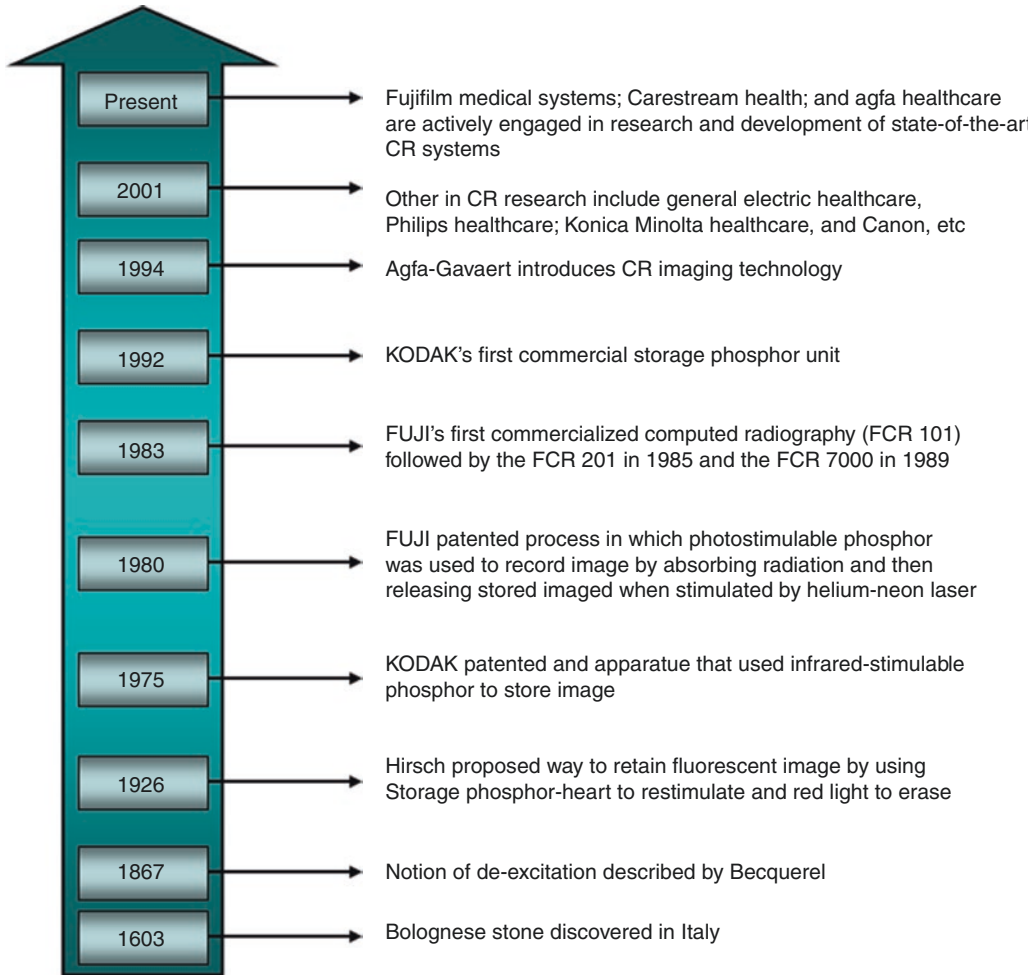

Fig. 3.1 Notable developments in the history of CR based on photostimulable luminescence. As can be seen, three manufacturers, most notably Agfa, Fuji, and Kodak (now Carestream), have been actively engaged in CR research and development

their FCR-101 unit. This was followed by other manufacturers, most notably Kodak and Agfa, whose systems became popular as well. Today other manufacturers are actively engaged in CR research, technology development, and marketing including Carestream (formerly Kodak), Agfa Healthcare, Konica Minolta Healthcare, and Canon, as well as other medical imaging vendors such as Philips Healthcare, Siemens Healthineers, and General Electric Healthcare. The interested reader may visit their respective websites for their most recent developments in hardware and software.

3.4 The CR Imaging System

The CR imaging system components are shown in Fig. 3.2. It is clear that the imaging process consists of four distinct steps: image acquisition, image plate scanning and erasure, image processing, and image display.

3.4.1 Image Acquisition

As illustrated in Fig. 3.2, image acquisition refers to X-ray exposure of the storage phosphor plate cassette or imaging plate (IP). It is at this point where the technologist must pay careful attention to technical details, such as positioning, centering of the X-ray beam, selection of the appropriate IP, grid selection, and correct radiographic exposure technique factors (kVp, mAs). Image acquisition also refers to the mechanism of X-ray interaction with the phosphor to produce a latent image and subsequent scanning of the IP by a laser beam to produce photostimulable luminescence (PSL). The scanning of the IP takes place in the imaging plate reader/processor. This readout process consists of essentially laser scanning, detection, and conversion of the PSL and digitization of the signal from the analog-to-digital converter (ADC). The IP is subsequently erased and can be used again for the next examination. These elements will be described in detail later in the chapter.

3.4.2 Image Processing

Image processing in CR refers to the use of several digital operations for pre-processing and post-processing of the CR image data. While pre-processing deals with shading corrections, pattern recognition, and exposure field recognition,

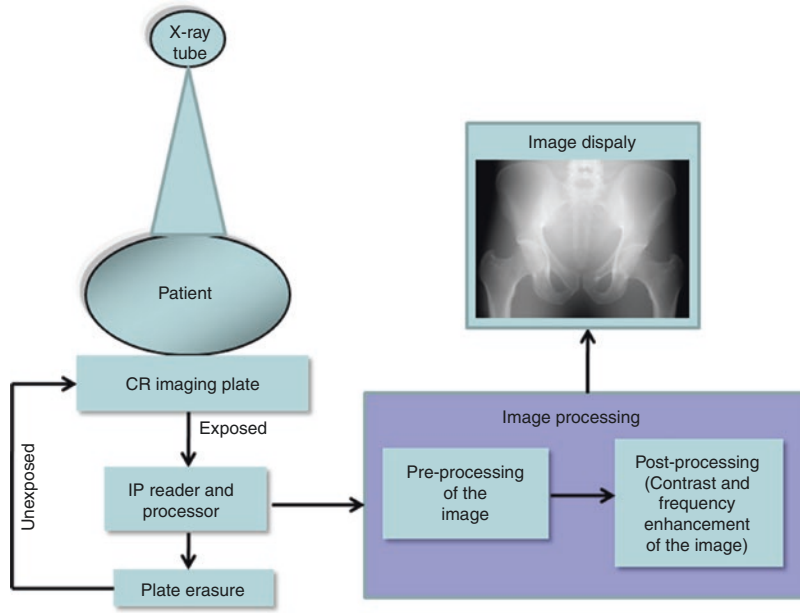

Fig. 3.2 The CR imaging system components. See text for further explanation

post-processing of the CR image data on the other hand refers to contrast enhancement (image grayscale processing) and edge enhancement, a technique which is based on frequency processing (Chap. 2). Additionally, energy subtraction imaging can be done in CR using a special algorithm to create separate images of bone and soft tissue. Image processing will be elaborated on in the section on image processing in this chapter.

3.4.3 Image Display, Storage, and Communications

After images have been processed, they are displayed for viewing and interpretation. In a *Picture Archiving and Communication Systems (PACS)* environment, the technologist determines and assesses the overall image quality of the image and subsequently sends the image to the PACS. Once in the PACS, images are retrieved for interpretation by a radiologist.

In CR, images are displayed on a computer workstation. These workstations have either a cathode-ray tube (CRT) display monitor or an active-matrix liquid crystal display (LCD) device. The workstation also allows the technologist to do any post-processing on images before they are communicated to the PACS for storage. Image storage and archiving in a CR-PACS environment include the use of magnetic tapes and disks, magneto-optical disks, optical disks, and digital videodisks (DVDs). Finally, communication of the CR image to the PACS is accomplished via computer networks.

3.5 Basic Physics of CR Image Formation

The CR imaging process, that is, the image acquisition in particular, consists of a three-step cycle of image formation, as illustrated in Fig. 3.3. The IP consists of a photostimulable storage phosphor (PSP) layered on a base to provide support. Photostimulable phosphors have the property of creating and storing a latent image, when exposed to X-rays. To render the latent image visible, the PSP must be scanned by a laser beam of a specific wavelength. Laser scanning produces a

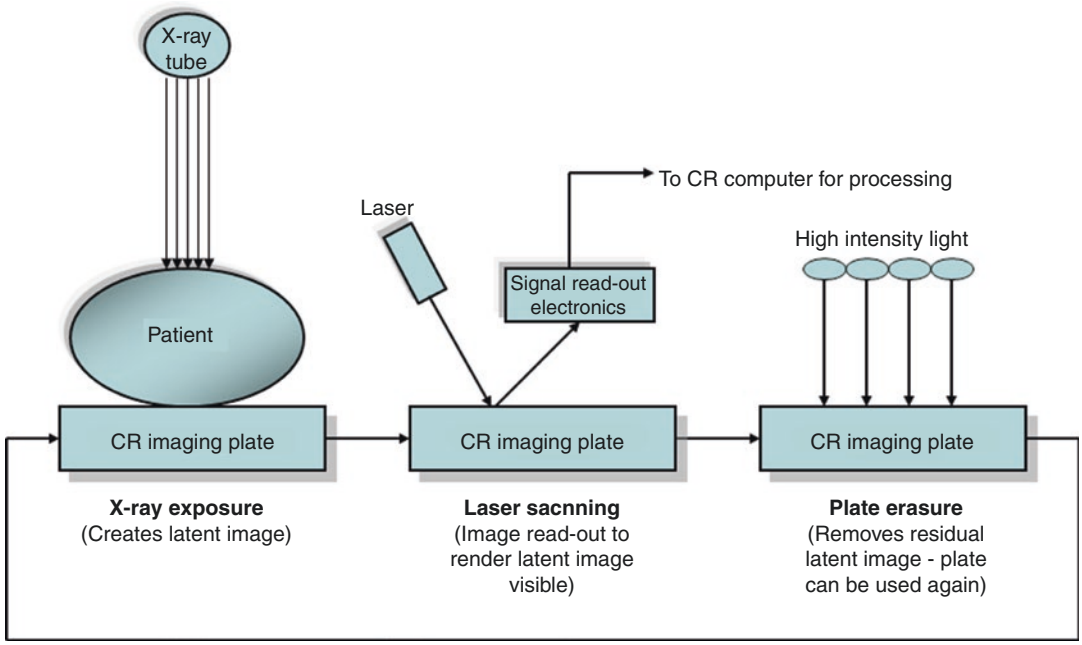

Fig. 3.3 The CR image acquisition process is a cycle consisting of three steps: X-ray exposure of the IP, laser scanning of the exposed IP, and erasure of the IP for subsequent reuse. See text for further explanation

luminescence (light) that is proportional to the stored latent image. This luminescence is referred to as photostimulated luminescence (PSL).

After laser scanning, the PSPIP is erased, by exposing it to a high-intensity light beam, to get rid of any residual latent image. This step is important so that the IP can be used again and again for several X-ray exposures. In this section, the basic physics of photostimulable storage phosphor X-ray exposure and photostimulated luminescence emission will be described.

3.5.1 Nature of PSPs

As can be seen in Fig. 3.3, the IP contains the PSP. The phosphors used in radiology must have certain physical characteristics to be useful in CR imaging. For example, phosphors should have good X-ray absorption efficiency and must be capable of being stimulated by a helium-neon (He-Ne) laser. Additionally, the luminescence light must be compatible with the photomultiplier tube (PMT) phosphor (for proper detection and capture), and the time for luminescence must be shorter than 1 μs [2]. Finally, these phosphors should be able to store the latent image for a number of hours without compromising the signal from the IP.

The phosphors that meet the above requirements and are used by several manufacturers are, in general, *barium fluorohalide: europium (BaFX: Eu^{2+})*. The halide (X) can be chlorine (Cl), bromine (Br), or iodine (I) or a mixture of them [3]. The phosphor is usually doped with Eu^{2+} which acts as an activator to improve the efficiency of PSL. Another phosphor used in CR is *BaFBr/I: Eu^{2+}*, and recently cesium bromide (CsBr: Eu^{2+}) is now used as a PSP for CR imaging [3].

As mentioned above, good X-ray absorption efficiency is one of the requirements of a PSP for CR imaging. Such efficiency depends on not only the kVp (X-ray energy) used but also the thickness of the phosphor used in the IP. The X-ray absorption efficiency of BaFBr PSP compared with the X-ray absorption efficiency of the rare-earth phosphors, gadolinium oxysulfide (Gd_2O_2S) and cesium iodide (CsI), is an important point to note. Between 35 keV and about 50 keV, BaFBr attenuates (absorbs) X-rays much better than Gd_2O_2S rare-earth screens because of the lower k-edge absorption of barium. Note, however, that at energies lower than 35 keV and greater than 50 keV, Gd_2O_2S attenuates much better than BaFBr [4].

3.5.2 Latent Image Formation and PSL

X-ray exposure of the PSPIP creates a latent image, and laser scanning of the exposed IP produces PSL. The information captured from the PSL is used to create the CR image. The physics of latent image creation and the mechanism of PSL are complex and beyond the scope of this book; however, the basic physics of image formation will be highlighted here. The interested reader should refer to the review article by Rowlands [2] for a detailed description of the physics of CR.

The mechanics of how PSPs is thought to work is shown in Fig. 3.4. When X-rays fall upon the PSPIP, the europium atoms are ionized by the radiation, and the electrons move from the valence band (ground state) to the conduction band (higher energy). Electrons in the conduction band are free to travel to a so-called F-center. F comes from the German *Farbe* meaning color [5]. The number of trapped electrons is proportional to the absorbed radiation. It is at the point in the process where the electrons are spatially distributed to create the latent image. In addition to this mechanism, X-ray exposure of the PSPIP causes it to fluoresce (emits light when it is exposed to X-rays) for a very brief duration.

To render the latent image visible, the PSPIP is taken to the CR reader/processor to be scanned by a laser beam. This process is referred to as photostimulated excitation Fuji [5]. While in the CR reader, the PSPIP is scanned systematically (to be described later in the chapter). The laser light used must be capable of being absorbed by the "F-centers." This absorption causes the trapped electrons to move up to the conduction

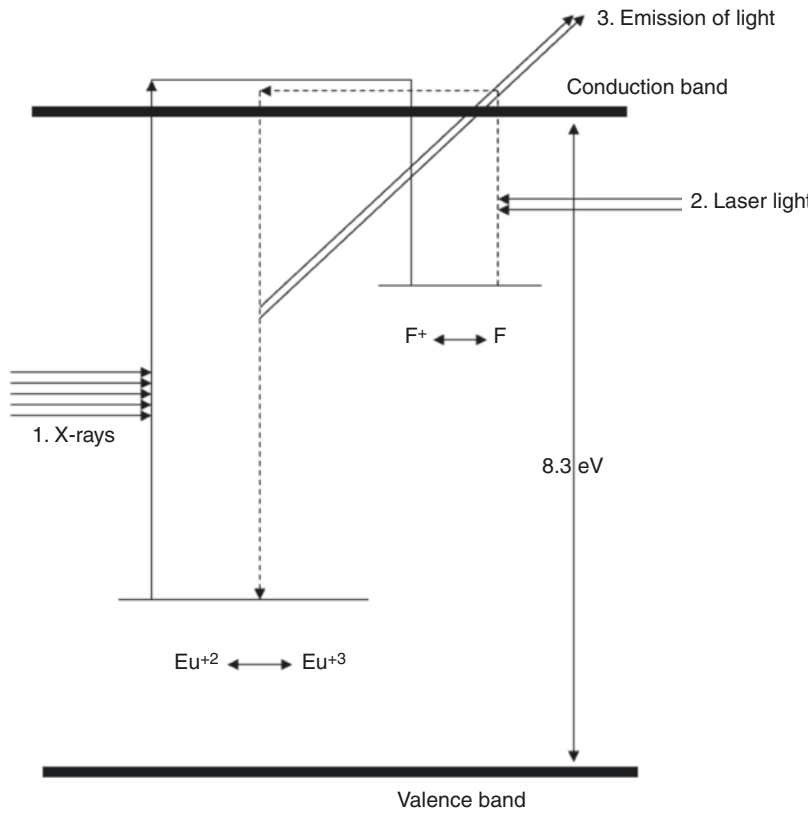

Fig. 3.4 The mechanism of latent image creation and photostimulable luminescence (PSL). See text for further explanation

band, where they are free to return to the valence band, thus causing the $Eu^{3+}$to return to the Eu^{2+} state. This transition of the electrons from a higher energy state to a lower energy state (ground state) results in an emission of bluish-purple light (~415 nm wavelength). This is referred to as *photostimulable luminescence (PSL)* in the PSPIP. This PSL is very different from the fluorescence described earlier. The lasers used today for PSL in CR units are semiconductor lasers that produce light with a 680 nm wavelength compared to He-Ne lasers that produce light with a 633 nm wavelength used in earlier CR units.

The PSL from the IP is collected by a special light collection device and sent to a photomultiplier tube that produces an electrical signal. This signal is subsequently digitized and sent to a digital computer for processing and CR image creation.

3.5.3 PSL Characteristics

There are several characteristics of PSL that are important in CR imaging. These include the light spectra related to the PSPIP and fading. The light spectra of the PSP include the storage phosphor stimulation and the PSL emission spectra. It is essential to note that the stimulation spectra are different from the PSL emission spectra. This is important to CR imaging since it is the PSL emission being captured by the PMT that creates the CR image.

Fading is a term that refers to the time it takes for the latent image to disappear. The latent image can last for several hours; however, it is important to read the exposed IP in a reasonable time, so as not to compromise the PSL signal. For example, the PSL decreases by about 25% if the time between exposure and image reading is 8 h [5].

3.6 CR Technology

The CR imaging system is shown in Fig. 3.2. The components of significance in this illustration are the imaging plate (IP) and the imaging plate reader/processor. Other components, such as image display and image processing, will be described later in the chapter.

3.6.1 The CR Imaging Plate

The digital detector used in CR imaging is the *imaging plate (IP)*. A cross section of the IP showing the major structural components is shown in Fig. 3.5. The IP consists of the PSP layer on a base that provides support. In addition, the IP structure consists of two protective layers: an electroconductive layer and a light-shielding layer. The support holds the other components of the IP together and provides mechanical strength. One of these critical components is the photostimulable phosphor. The phosphor is mixed with an organic binder (e.g., polymer, such as polyester) and coated onto the support layer. Two layers coat the phosphor, a front and a back protective layer (Fig. 3.5). The front protective layer must be constructed so that it provides durability during multiple uses. This layer must also allow light from the laser and the stimulated light to pass through it. The purpose of the electroconductive layer, shown in Fig. 3.5, is to reduce any problems when the IP is transported in the CR reader (CR scanner or processor) and static electricity problems that may degrade image quality. Some IPs have a barcode for easy identification.

Essentially, there are two types of IPs: a *standard-resolution IP* and a *high-resolution IP*. While standard-resolution IPs have thick phosphor layers and absorb more radiation, high-resolution IPs have thinner phosphor layers and provide sharper images compared to thick phosphors. The sharpness is due to the fact that thinner phosphors reduce the lateral spread of the laser light.

In terms of radiographic speed, thick phosphor IPs have faster speeds than high-resolution IPs (slow speeds), similar to the cassettes used in conventional radiography. Of course, the high-resolution IPs will be used for extremity imaging and other small parts where detail (sharpness) is critical. The imaging plate size varies depending upon the manufacturer; however, 17″ × 17″ (43 × 43 cm), 17″ × 14″ (43 × 35 cm), 14″ × 17″ (35 × 43 cm), and 14″ × 14″ (35 × 35 cm) are not uncommon. Smaller sizes are also available. CR IPs are housed in cassettes similar to conventional film-screen cassettes, with the IP replacing the film, and there are no intensifying screens. The CR cassette is usually made of aluminum (Fujifilm) or aluminum honeycomb panel (Carestream).While the front of the cassette is radiolucent, the back of the cassette is designed with a lead backing to prevent backscatter

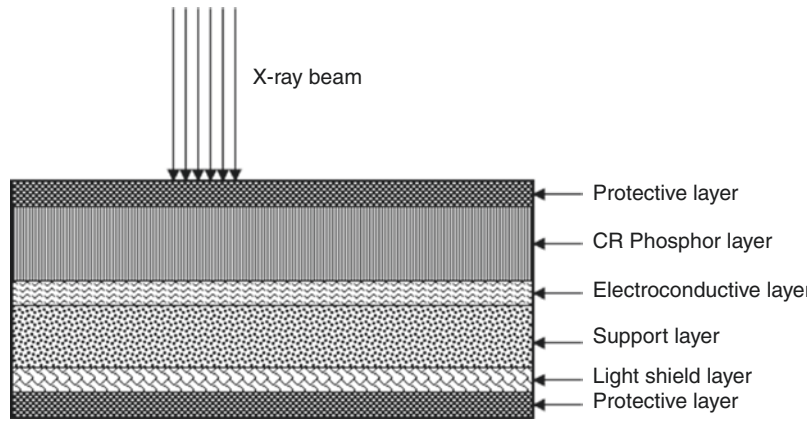

Fig. 3.5 A cross section of a typical imaging plate (IP) used in CR imaging, showing the major structural components

radiation from getting to the IP. Backscatter will lead to image artifacts.

3.6.2 The IP Imaging Cycle

One of the advantages of CR is that the IP can be used over and over again for several hundreds of exposures. During imaging, the IP goes through the imaging cycle shown in Fig. 3.2. This cycle basically consists of at least three steps: X-ray exposure of the IP, readout of the exposed IP, and erasure of the IP. In the first step, the "ready-to-use" IP is exposed to X-rays using radiographic exposure factors (kVp, mAs) suitable to the needs of the examination. X-ray exposure produces an immediate light emission but also creates a latent image in the form of energy storage in the phosphor. The energy stored is directly proportional to the intensity of X-rays striking the phosphor. The exposed IP is then readout in the CR reader (CR processor) to render the latent image visible, while in the CR reader, a laser light scans the IP to produce the photostimulated luminescence as described earlier. The mechanics and electronics of this readout procedure will be described in the next section. Finally, in the third step of the IP imaging cycle, the IP is erased using a high-intensity light to remove any residual energy after the IP has been scanned by the laser beam. The erased IP is now ready to be used again.

3.6.3 The CR Reader: Types

The *CR reader*, or *scanner*, as it is sometimes referred to, is a machine for scanning the exposed IP to render the latent image visible. To do this efficiently and effectively requires a set of technical components engineered precisely to allow the IP to be scanned, erased, and made available for repeated use by the technologist. The purpose of the CR reader is to render the latent image stored on the exposed IP visible. The electrical signal generated as a result of scanning the IP is amplified and subsequently digitized.

There are two types of CR systems: *cassette-based systems* and *cassetteless systems*.

3.6.3.1 Cassette-Based Systems

These systems use individual IPs of different sizes analogous to cassette-based film-screen radiography. As seen in Fig. 3.2, an exposed IP has to be physically taken to a CR reader unit for scanning to acquire an image from the IP, followed by erasure of the IP and subsequent transport of the IP in the unit itself for reuse. This means that the IP is in contact with various transport mechanisms that may in time result in plate/phosphor damage

3.6.3.2 Cassetteless Systems

These systems evolved to overcome some of the problems with cassette-based systems. One such problem, for example, is related to physical task of taking an exposed IP to the CR reader for processing, as well as mechanical transport of the IP itself in the reader. Cassetteless CR systems incorporate a fixed stationary single IP that is encased in a special housing that forms a part of the unit. There is also no contact with the IP in the unit when it is read. The single fixed IP can accommodate various exposure sizes ranging from 17″ × 17″ and 14″ × 14″ to 10″ × 12″ and 8″ × 10″ or 10″ × 8″. These varying sizes will have varying matrix sizes as well.

Once the fixed stationary IP is exposed, the patent image is acquired using a scanning technology appropriate to the system.

3.6.4 The CR Reader: Scanning Technologies

Acquiring the image from the exposed IP can be accomplished by *point-scan (P-S)* CR readers or by *line-scan* CR readers. The major components of a P-S CR reader are shown in Fig. 3.6 and include the laser source, the IP transport mechanism, light channeling guide, photodetector (PMT), and the analog-to-digital converter (ADC). Each of these will now be described briefly.

Firstly, the IP is removed from the cassette and is placed on the transport mechanism for scanning by a *laser beam*. While the movement of the IP is referred to as the "slow-scan"

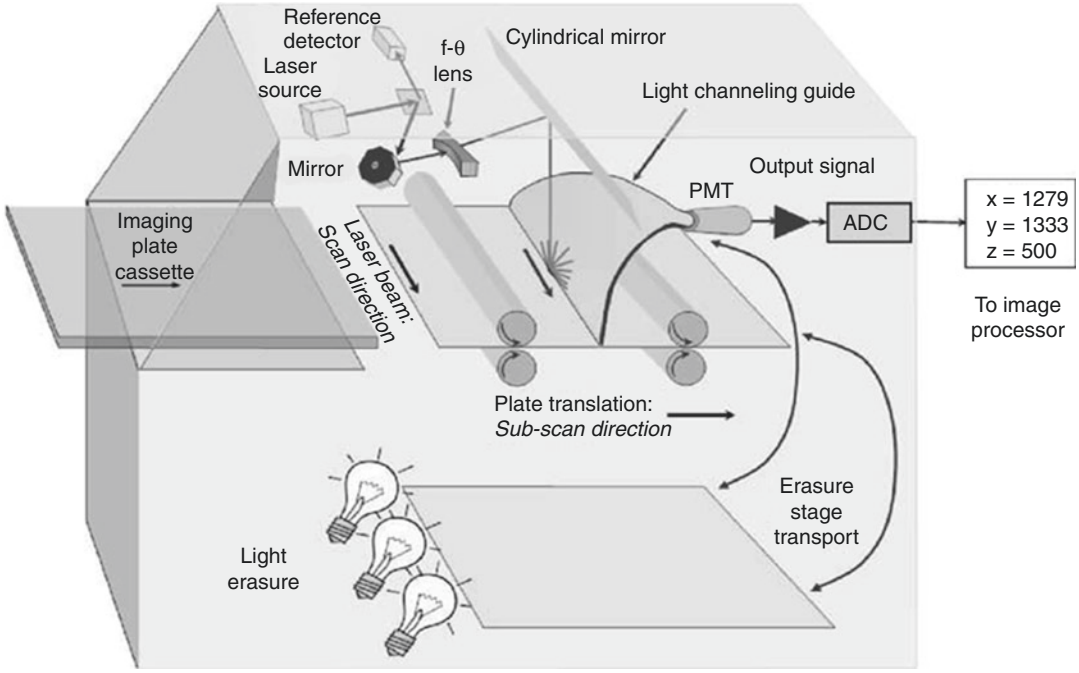

Fig. 3.6 Major components of a typical PSP reader showing how the IP is scanned. A stimulating laser source, a beam splitter, oscillating beam deflector, lens, cylindrical reflecting mirror, light collection guide, photomultiplier tube (PMT), and light erasure stage. The IP is erased to remove all residual signals and returned to be loaded in the cassette (From AAPM Report 93: Acceptance Testing and Quality Control of Photostimulable Storage Phosphor Imaging Systems-October 2006. Reproduced by permission of the AAPM)

direction, the laser beam movement across the IP is called the "fast-scan" direction [4]. The laser beam is used to stimulate the trapped electrons (latent image) in the exposed IP. In the past, gas lasers such as a helium-neon (He-Ne) laser were used; however, current CR readers use solid-state laser diodes that emit a red laser beam having wavelengths of about 670–690 nm. The laser stimulation of the IP causes it to emit light (by photostimulable luminescence (PSL) that is of a much different wavelength than the stimulating laser light.

Secondly, the emitted light from the IP is optically filtered and collected by the light channeling guide or light collection optics as it is sometimes referred to as. This light (PSL) is then sent to the photodetector (a photomultiplier tube) or charge-coupled device (CCD) which converts the PSL into an electrical signal (analog signal) that is first amplified and subsequently digitized by the ADC. It is not within the scope of this book to describe the mechanics of amplification. Finally, the analog signal from the photodetector is sent to the ADC for digitization as described in Chap. 2. Digitization involves both sampling the analog signal and quantization. Depending on the amplification, the ADC will produce 8–16 bits of quantization per pixel, providing discrete gray levels ranging from 2^8 to 2^{16}.

CR systems use several linear laser sources, a lens system, and a linear array of CCD photodetectors. While the laser beam is collected and shaped by the lens system, to scan the IP line by line (instead of point by point for point-scan systems), the PSL from the IP is collected by the CCD linear photodetector array. In this manner, the line-scan CR reader is much faster than the point-scan CR reader.

As described above, the IP is scanned on one side to emit the PSL that is used to produce an image. Recently, however, dual-sided reading technology has become available in more recent

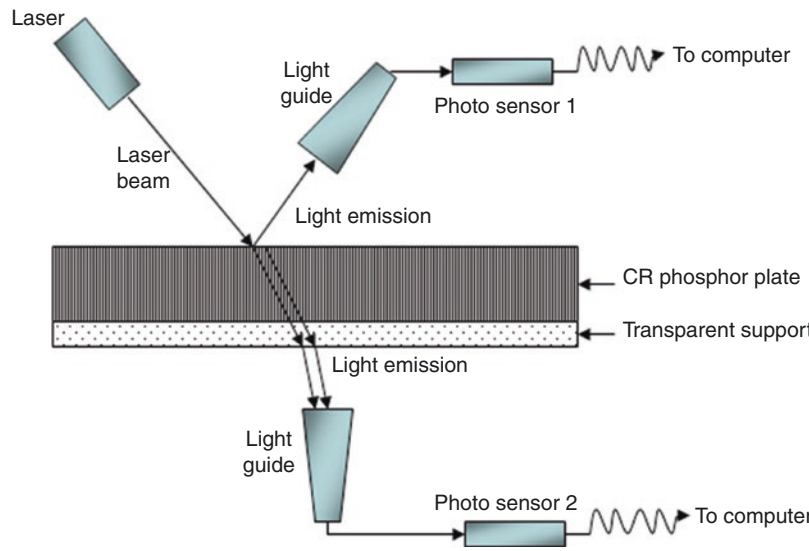

Fig. 3.7 The mechanism of dual-side reading of the IP in CR imaging

CR units. *Dual-sided reading* of the IP is illustrated in Fig. 3.7. It is apparent that two sets of photodetectors (dual light collection system) are used to capture PSL from the front and back side of the IP. In this way, more signal is obtained to improve the signal-to-noise ratio and, hence, improve image quality. Additionally, a thicker phosphor layer can be used to increase the absorption of X-rays, hence improving system efficiency.

After the latent image has been extracted by laser stimulation and collection of the PSL from the IP, the IP must be erased to prepare it for another exposure. *Erasure of the IP* is done in the CR reader by exposing it to a high-intensity light that is brighter than the stimulating laser light, to get rid of any residual signal left on the IP. It is interesting to note also that since the IP is also sensitive to background radiation, as well as scattered radiation from X-ray procedures, it must be erased before use, especially if the IP has not been used for a period of time.

3.6.5 The CR Workstation

The *CR workstation* provides the technologist with the opportunity to interact with the entire CR process by facilitating a number of important functions ranging from the input of patient identification or selection of patient data and target exposure and image preview, image processing, quality assurance procedures, image printing, and sending images to the PACS through the Digital Imaging and Communication in Medicine (DICOM) standard.

A typical CR workstation consists of the image processing computer and image display monitor, keyboard, and mouse. In addition, some workstations offer a barcode reader and a magnetic card reader (not shown). While cathode-ray tube (CRT) monitors are used for some CR workstations, liquid crystal display (LCD) monitors have become commonplace. These LCD monitors are available in different sizes; however, 21-inch and 20-inch monochrome monitors are not uncommon. A 3-megapixel monitor offers high-resolution display of pixels with a bit depth of 8 ($2^8 = 256$). With high-resolution monitors, radiologists can perform soft-copy reading of CR images.

The keyboard allows the technologist to input relevant information about the CR examination, and the mouse allows the selection of various operations. Some monitors utilize touch-screen technology to enable the technologist to communicate with the software. Finally, the barcode reader enables the registration of the various IPs used.

In general, CR software should be intuitive to provide ease of use of the system. Such software will allow the technologist to perform several functions including quality assurance tasks and simple and complex image processing operations. Image processing will be described subsequently.

3.6.6 Computer Networking and CR

Once images are acquired and displayed for viewing by the technologist, they must be assessed for image quality and QA before they are sent to the PACS. Additionally, the CR system is interfaced to the hospital information system (HIS) and the radiology information system (RIS). These components, that is, the CR unit (CR reader and workstation), the HIS/RIS, and the PACS, should be fully integrated to communicate with each other. While DICOM facilitates the communication of images in the digital radiology environment, the Health Level-7 (HL-7) standard addresses the communication of textual data within the information systems environment. PACS and information systems, computer networking, and communications standards such as DICOM and HL-7 will be explored further in Chap. 10.

3.7 Digital Image Processing in CR

The overall general concepts of *digital image processing* were described in Chap. 2. Image processing operations, such as image contrast and brightness control as well as spatial frequency filtering while images can be enhanced for sharpness and blurred for the purpose of meeting the viewing needs of the observer, were reviewed.

Image processing in CR can be discussed in terms of pre-processing and post-processing operations both of which are intended to enhance the visual appearance of the image displayed for viewing on a monitor (soft-copy viewing), in an effort to assist the radiologist in image interpretation.

3.7.1 Pre-processing Operations

There are several *pre-processing operations* used to identify, correct, and scale the raw image data obtained when the IP is scanned in the CR reader and before the image is displayed for viewing and subject to post-processing. Pre-processing operations are also referred to as acquisition processing.

Pre-processing operations for digital detectors are several, and, more importantly, each manufacturer offers proprietary algorithms for their systems. Therefore, it is not within the scope of this book to describe these algorithms; however, there are a few noteworthy aspects of pre-processing that the technologist must be familiar with and will be reviewed here. Pre-processing in CR is essential to correct the raw digital data collected from the IP and the CR reader that may have imperfections. For example, the IP may have scratches and other marks, and the CR reader may have dirt on the light channeling guide, all of which will lead to image artifacts. In addition, the raw data is scaled to ensure that only the useful anatomic signals are used in the digitization process to improve image quality.

One important pre-processing method in CR is *exposure field recognition* also referred to as exposure data recognition (Fujifilm Medical Systems) and segmentation (Carestream). The basic steps of exposure recognition are shown in Fig. 3.8 and include exposure field recognition, histogram analysis, and grayscale rendition. The purpose of exposure recognition is to identify the appropriate raw data values (minimum and maximum values) to be used for image grayscale rendition and to provide an indication of the average radiation exposure to the IP CR detector. The latter will be described later in this chapter.

In the first step of exposure field recognition, the collimation edges or boundaries are detected, and anatomical structures that should be displayed in the image are identified using specific algorithms, such as the "shift and subtract" method where "the image is subtracted from an identical copy of itself and then shifted onto horizontal and vertical directions by two or more

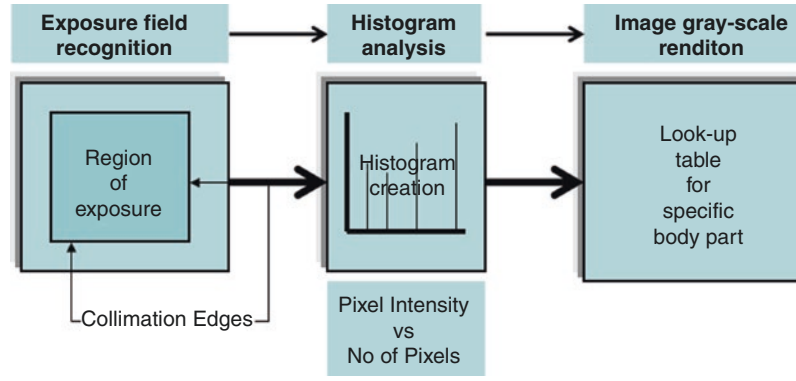

Fig. 3.8 The basic steps of exposure recognition are exposure field recognition, histogram creation, and grayscale rendition. The histogram in this case is referred to as the scanned or measured histogram to distinguish it from the stored histogram

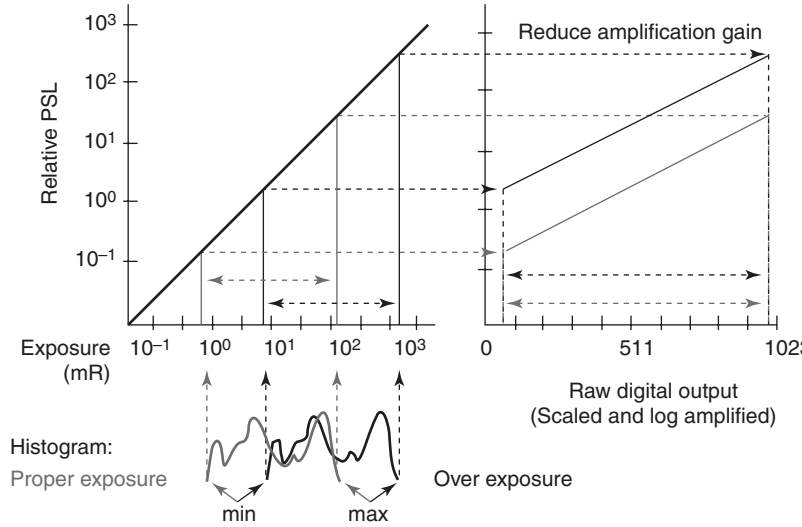

Fig. 3.9 Digital systems compensate for under- or overexposure by automatically adjusting the internal gain of the output conversion stage once the minimum and maximum values are identified (see overexposure histogram (dotted line)). The histogram shifts position but does not change shape. Although not shown, the digital system gain curve can also adjust slope to adjust for variation of the input exposure dynamic range (PSL, photostimulable luminescence). (From AAPM Report 93: Acceptance Testing and Quality Control of Photostimulable Storage Phosphor Imaging Systems-October 2006. Reproduced by permission of the AAPM)

pixels. This produces differential signals at locations of rapid change (e.g., collimator shadows) and identifies the area of interest" [4]. In the second step, a *histogram* of the information on the IP (including the anatomy and the collimated and opened regions of the image) is created. This is referred to as the *measured histogram* or scanned histogram, to distinguish it from what is referred to as known or *stored histograms*, that is, the identical copy of the images of the anatomy under study stored previously in the machine. The CR imaging system will compensate for underexposure or overexposure by matching the measured histogram with the appropriate known histogram using anatomy-specific template matching algorithm (Seibert [4]). This is illustrated in Fig. 3.9 where the measured histogram from an overexposure (dotted line) is shifted to match the known histogram (solid line) stored in the machine to produce an acceptable image. This is accomplished by rescaling that "involves mapping the minimum useful value to a correspondingly small digital value and mapping the maximum useful value to a correspondingly large

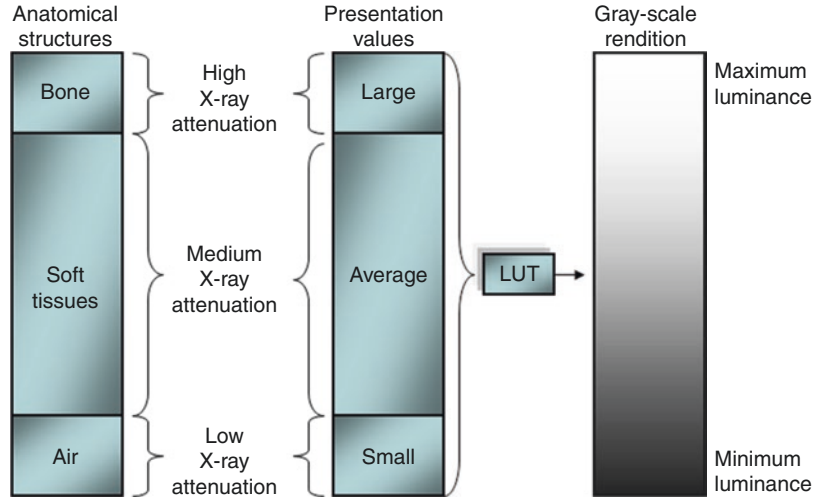

Fig. 3.10 Grayscale rendition maps the degree of penetration of the anatomical regions to the degree of luminance of the display device. Large and small penetration values are mapped to the maximum luminance and minimum luminance, respectively

digital value within the typical output image (10–12 bits)."

The final step in exposure recognition is *grayscale rendition* of the image. Grayscale rendition is a procedure that "maps the raw image values for the least penetrated anatomic region to the largest presentation value for display at maximum luminance. The most penetrated anatomic region of interest is mapped to the smallest presentation value for display at minimum luminance. The intermediate raw values are then mapped to presentation values in a monotonically decreasing fashion. This produces a presentation with a black background similar to that of conventional radiographs" [6]. This is illustrated in Fig. 3.10. Note that a LUT is used to do this function. Exposure recognition may not always be successful and may fail due to problems with too much scattered radiation, metallic components in the patient such as implants, the presence of lead markers, and immobilization devices [6].

In CR, exposure recognition performs another task, and that is to provide an indication as to the amount of radiation falling upon the detector as a result of the exposure technique used by the technologist. This *exposure indicator*, or *exposure index* as it is sometimes referred to as, appears on the displayed image and serves to provide the technologist with a visual cue as to whether correct or incorrect exposure technique was used. Incorrect exposure techniques are those that result in underexposure or overexposure of the patient. In this regard, the exposure indicator can be used as a quality control tool to facilitate the optimization of radiation protection.

Manufacturers of CR systems report the exposure indicator in different ways, and at present there is no universal standard for reporting exposure indicators. For example, while Fuji Medical Systems refers to their exposure indicator as a *sensitivity (S) number* (S-number), Agfa Medical Systems uses the *log of median (lgM)* value of the histogram, and Carestream uses the term *exposure index (EI)*. Exposure indicators will be described in more detail later in this chapter.

3.7.2 Post-processing Operations

Post-processing of the displayed image, to suit the viewing needs of the observer, who may want to not only sharpen and reduce the noise in the image but also enhance the image contrast as well, logically follows pre-processing. There are several types of *post-processing* algorithms for use in CR. These in general include contrast enhancement; spatial frequency or edge enhancement; multi-scale, multifrequency enhancement; and dual-energy and disease-specific processing.

Contrast enhancement is also referred to as contrast scaling, and various manufacturers use different terms to refer to it. For example, while

Fuji uses the term gradation processing, Carestream and Agfa use the terms tone scaling and latitude reduction, respectively. A more current algorithm is one from Agfa called multi-scale image contrast amplification (MUSICA).

The purpose of contrast enhancement is to optimize the image contrast and density to enhance diagnostic interpretation of the image. Essentially the pixel values are normalized and rescaled using a Look-Up Table (LUT). An example of a very common approach for contrast enhancement is the unsharp mask method described in Chap. 2.

Another common post-processing operation is *edge enhancement* or *spatial frequency processing*. These algorithms are intended to adjust or control the sharpness or detail of an image by adjusting the frequency components of the image. As described in Chap. 2, an image in the spatial location domain can be transformed into an image in the frequency domain using the Fourier transform (FT). The spatial frequency domain image contains both high spatial frequencies (detail information) and low spatial frequencies (contrast information). The effect of frequency processing on an image of the author is shown in Fig. 3.11.

More recent algorithms (software) for frequency processing have been introduced by CR vendors, who use their own terminology to describe their software. For example, while Fujifilm Medical Systems uses the term multi-objective frequency processing (MFP), Agfa Healthcare and Carestream use the terms multi-scale image contrast amplification (MUSICA) and enhanced visualization processing (EVP), respectively. In addition, Philips Healthcare and Konica Minolta Healthcare have algorithms called Unified Image Quality Enhancement (UNIQUE) and Hybrid Processing, respectively. While UNIQUE is a multi-resolution algorithm, the Hybrid algorithm uses distinct frequency components of different anatomical parts on the image, to reduce noise and shadowing resulting in a more natural appearance of the image while maintaining detail. CR manufacturers also offer a wide variety of image processing software for their systems. While it is not realistic to describe them all in this text, a few advanced image processing options for one manufacturer (Fuji) are highlighted in Table 3.1.

It is important that technologists realize that the information content of an image depends primarily on the radiation dose used. During the

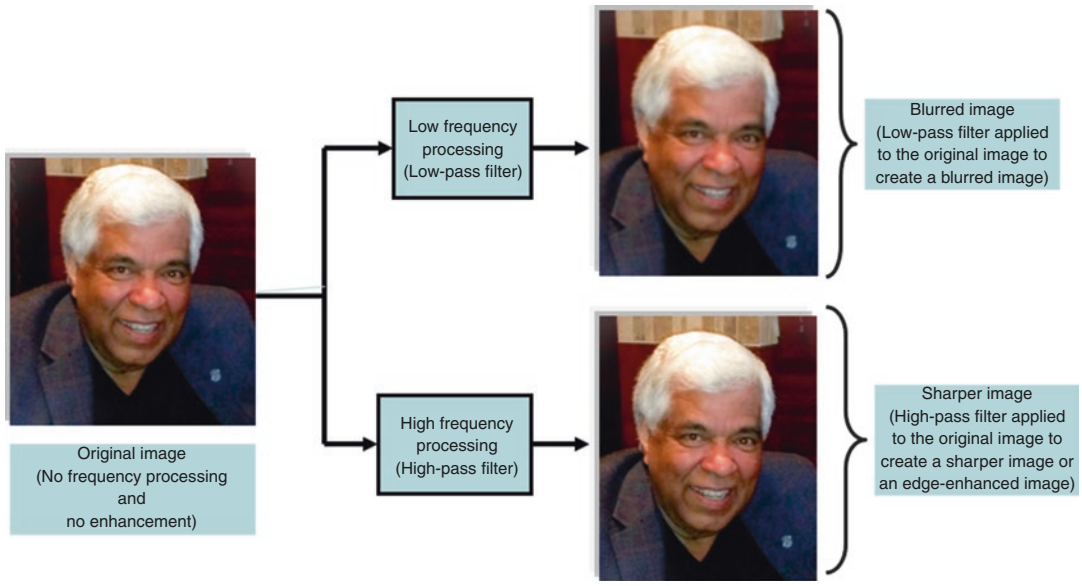

Fig. 3.11 The effects of low- and high-frequency processing (edge enhancement) on the visual quality of a photograph of the author, Euclid Seeram

Table 3.1 Several examples of Fuji's advanced image processing software for their CR imaging systems (From Fuji's Image Intelligence Brochure)

Image processing operation	Brief description
Dynamic range control (DRC)	Improves the visualization of areas with different densities in the same image
Energy subtraction	Using a single exposure of a body part, three different displays such as bone, soft tissue, and the standard exam (chest in this case) to allow visualization and typification of structures that may be obscured by overlying or underlying anatomy
Multi-objective frequency processing (MFP)	This is frequency enhancement which adjusts both large and small structures independently within the same image simultaneously
Flexible noise control (FNC)	This software separates noise and image signals, enabling selective suppression of noise levels to enhance diagnostic interpretation
Grid pattern removal (GPR)	Removes grid patterns from the image to suppress moiré patterns within an image
Image composition	Separate images are stitched (joined together) to form a single image

conduction of an examination, the technologist must strive to use the appropriate exposure technique factors (kVp and mAs) that produce the best possible image quality, with the least radiation dose to the patient. This task requires a careful assessment of all the parameters that influence the selection of appropriate exposure technique factors. The technologist should not depend on the use of inappropriate image processing to improve image quality. The proper use of image processing can improve image quality, and this should be the primary goal of the technologist who is involved in the use of these image processing algorithms to enhance digital images for diagnostic interpretation.

3.8 Exposure Control in CR

The use of proper exposure technique factors is a vital part of any radiographic examination, since these factors determine the image quality obtained as well as the radiation dose to the patient.

3.8.1 IP Response to Exposure

The response of radiographic film to radiation exposure is well understood as is described by the characteristic curve or the H and D curve as it is often referred to (Chap. 1). When the exposure is low, the film is underexposed and the image is light and is not acceptable. When the exposure is high, the film is overexposed and the image is dark and is not acceptable. In both cases repeat exposures are required to achieve the proper film density using image receptor exposures that fall within the slope of the characteristic curve. This slope defines the useful range of exposures referred to as the film latitude or dynamic range of the image receptor.

Underexposure of the film results in noisy images while overexposure will result in high doses to the patient.

The above problems are solved by a CR imaging system since the IP has wider exposure latitude than film, as shown in Fig. 3.12. The consequence of this wide dynamic range on the image quality is also clearly illustrated in Fig. 3.12, which shows one of the significant advantages of CR. If the exposure is too low or too high, the image quality is still acceptable due to the ability of the CR system to perform digital image processing to adjust the image quality to match the image quality that would be produced by the optimum exposure. As seen in Fig. 3.12, a low exposure (underexposure) will produce high noise (that can be detected by the radiologist), while a high exposure (overexposure) will produce very good images, compared to the optimum image produced by the optimum exposure (appropriate exposure).

The fundamental problem with high exposures is that of increased radiation dose to the patient. As noted by Dr. A Seibert, "because of the negative feedback due to underexposures, a predictable and unfortunate use of higher exposures, '*dose creep*', is a typical occurrence. To identify an estimate of the exposure used for a

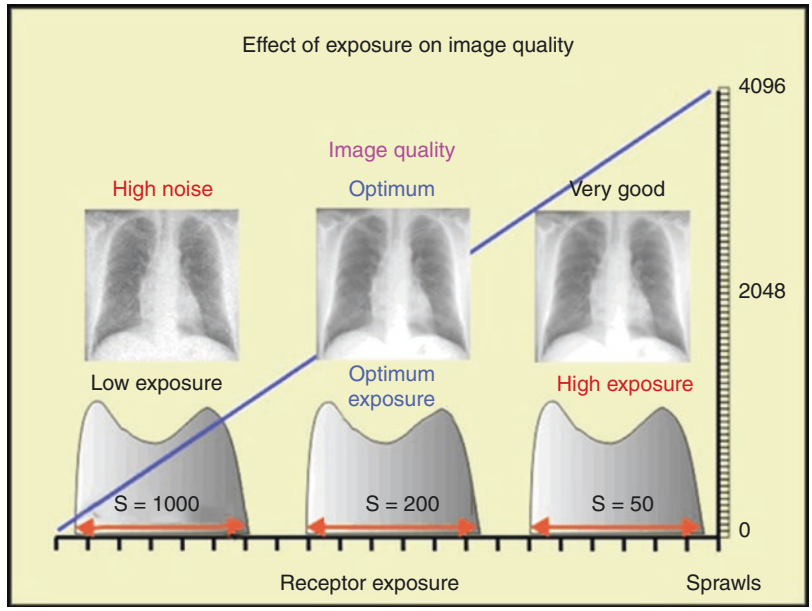

Fig. 3.12 CR has wide exposure latitude compared to film-screen radiography. The advantage of this is that CR can produce images that appear visually the same. See text for further explanation (Courtesy of Dr. Perry Sprawls, Emory University)

given image, CR manufacturers have devised methods to analyze the digital numbers in the image based upon the calibrated response to known incident exposure" [4]. As noted earlier in this chapter, one of the functions of exposure data recognition is to provide an indication of the amount of radiation falling upon the CR IP. This is referred to as the exposure indicator or exposure index.

3.8.2 Exposure Indicators

An *exposure indicator* is a numerical parameter used to monitor the radiation exposure to the IP in CR imaging. The determination of exposure indicators differs among CR vendors. As noted earlier, while Fujifilm Medical Systems refers to their exposure indicator as a sensitivity number (*S*-number), Carestream uses the term exposure index (EI). Agfa Healthcare, on the other hand, uses the term log of the median of the image histogram (LgM, *log of median* values); and Konica Minolta Healthcare uses the term sensitivity value (*S*-value). Only the first three will be described briefly in this chapter.

3.8.2.1 Sensitivity Number (*S*-Number) of the Fujifilm Medical Systems CR System

Fuji's *sensitivity number (S)* is derived from the following relationship, using a standard-resolution IP and under normal processing operations [4].

$$S = \frac{200}{\text{Exposure to the IP}\left(\text{mR}\right)}$$

This relationship shows that *S* is inversely proportional to the exposure; hence a low exposure will result in a high *S*-number, and a high exposure will result in a low *S*-number. It should be noted that if the exposure to the IP is low, the PMT signal will be weak and must be increased. On the other hand, if the exposure to the IP is high, the PMT signal is strong and must be decreased to produce optimum image quality. Adjusting the PMT's signal in this manner means that the sensitivity of the PMT will be set for the final scan of the IP. Recall that the IP undergoes a first scan to generate a histogram based on the collimation boundaries. This histogram is labeled the scanned histogram to distinguish it from stored histograms (pre-programmed histograms

for various body parts to be imaged). During the final scan or the final read of the IP, the scanned histogram is compared with the pre-programmed histogram.

This adjustment of the sensitivity of the PMT shows the amount of amplification that the system applies to the image signal to produce the proper image in order to match the scanned histogram to the correct pre-stored histogram for the body part under investigation.

The relationship S = 200/exposure (mR) means that an exposure of 1mR {0.258×10^{-3} millicoulombs/kilogram (mC/kg)}, a known incident exposure to the IP, will result in an S-number of 200. A low exposure of 0.1mR (0.258×10^{-4} mC/kg) will result in an S-number of 2000, while a high exposure of 10mR (0.258×10^{-2} mC/kg) will result in S-number of 20. Fig. 3.12 shows the effect of S-numbers of 1000 (low exposure), 200 (optimum exposure), and 50 (high exposure) on image quality. The challenge imposed by the CR imaging system is for the technologist to optimize the exposure indicator (in this case, the S-number), to reduce the dose to the patient while maintaining optimum image quality, so as not to compromise the diagnostic interpretation of the image.

The S-number can be thought of as being equivalent to the speed of the IP. If the exposure is low, the speed is increased (hence the S-number is large, say S = 1000) as shown in Fig. 3.12 and the image will be noisy. If the exposure is high, the speed will be decreased (low S-number, say S = 50, as shown in Fig. 3.12), and the image is very good, but at the expense of higher dose to the patient.

3.8.2.2 Exposure Index (EI) of the Carestream CR System

Carestream's *exposure index (EI)* also provides information about the average exposure to the IP, and it is calculated using the following relationship:

$$\mathrm{EI} = \log\left(\text{Exposure in mR}\right) \times 1000 + 2000$$

This relationship shows that the EI is directly proportional to the exposure; hence a high exposure will result in a high EI and a low exposure will generate in a low EI. The relationship also shows that a 1mR exposure (under certain conditions) will produce an EI of 2000. Exposures of 0.1mR (low) and 10mR (high) will result in EIs of 1000 and 3000, respectively. If the exposure is doubled, the EI will increase by 300. On the other hand, a reduction of the exposure by 0.5 (one-half) will reduce the EI by 300 [4].

3.8.2.3 Log of Median (LgM) Values of the Agfa Healthcare CR System

The Agfa Healthcare CR system uses an exposure indicator called the *log of median (LgM)* that is related logarithmically to the median of the IP exposure. Under certain conditions, the relationship is given as:

$$\mathrm{LgM} = 2.2 + \log\left(\text{Exposure in mR}\right)$$

If the exposure to the IP is 1mR, then the LgM would be 2.2.

3.8.3 Exposure Indicator Guidelines

Exposure indicators can be used as a means of monitoring the dose to patients while maintaining acceptable image quality. This task is an essential component on a quality assurance/quality control (QA/QC) program. The exposure indicator depends on a number of factors, for example, the kVp, filtration, patient positioning, source-to-image receptor distance (SID), collimation, beam centering, image processing algorithms, and so forth. Therefore guidelines for the QC of the exposure indicator are necessary. It is important for users to refer to the specific vendor for such guidelines.

3.8.4 Standardized Exposure Indicator

The differences in the exposure indicators described above have created not only confusion but also frustration among users. This has

provided the motivation for the development of a standardized EI, championed most notably by the International Electrotechnical Commission (IEC) [7] and the AAPM [8]. The nature and characteristics of the international standardized EI for digital radiography, through a description of standardization efforts, deviation index (DI), and the target EI (EI_T), and responsibilities of both the manufacturers and users will be covered in Chap. 5. An insight into optimization research will also be presented as a means of illustration of a dose-image quality optimization strategy to objectively determine EI_T values

3.9 Image Quality Descriptors

The *image quality descriptors* for a digital image such as a CR image are illustrated in Fig. 3.13 and include spatial resolution (detail), density resolution, noise, quantum detective efficiency (DQE), and artifacts. This chapter will not describe these in any detail; however, the essential elements of each will be highlighted.

3.9.1 Spatial Resolution

The *spatial resolution* of a digital image is related to the size of the pixels in the image matrix. Different sizes of IPs have different pixel sizes. For example, IP sizes of 35 cm × 43 cm (24″ × 17″), 23 cm × 30 cm (10″ × 12″), and 18 cm × 24 cm (8″ × 10″) for one manufacturer have pixel sizes of 0.2 mm, 0.14 mm, and 0.1 mm, respectively. The smaller the pixel size, the better the spatial resolution of the image.

The *pixel size* (PS) can be calculated using the relationship:

$$\text{PS} = \text{Field of view}\left(\text{FOV}\right) / \text{matrix size}$$

Thus, for the same FOV, the greater the matrix size, the smaller the pixels and the better the image sharpness. Typical CR image matrix size is 2048 × 2048. There are other factors affecting the sharpness of the digital image, but they will not be described here.

3.9.2 Density Resolution

The *density resolution* of a digital image is linked to the *bit depth*, which is the range of gray levels per pixel (see Chap. 2). An image with a bit depth of 8 will have 256 (2^8) shades of gray per pixel. In general, the greater the bit depth, the better the density resolution of the image.

3.9.3 Noise

Noise on the other hand can be discussed in terms of electronic noise (system noise) and quantum noise (quantum mottle). The quantum noise is determined by the number of X-ray photons (often referred to as the signal, S) falling upon the detector to create the image. While low exposure factors (kVp and mAs) will produce few photons at the detector (less signal and more noise, N),

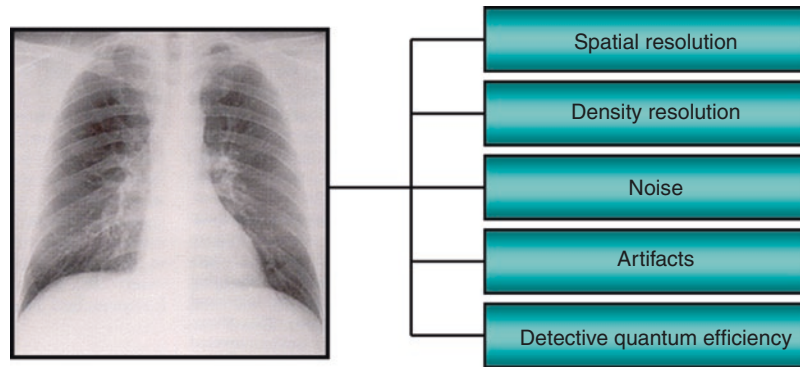

Fig. 3.13 Five image quality descriptors for a digital image such as a CR image

higher exposure factors will generate more photons at the detector (more signal and less noise). The former will result in a noisy image (grainy image) that is generally a poor image, and the latter will produce a better image at the expense of increased dose to the patient. The noise increases as the detector exposure decreases.

3.9.4 Detective Quantum Efficiency

The final descriptor of digital image quality is the *detective quantum efficiency* (DQE). The overall concept of the DQE for a digital detector is illustrated in Fig. 3.14. As can be seen, the detector receives an input exposure (incident quanta) and converts it into a useful output image. The DQE is a measure of the efficiency and fidelity with which the detector can perform this task. Note that the DQE also takes into consideration not only the signal-to-noise ratio (SNR) but also the system noise and therefore includes a measure of the amount of noise added.

The DQE can be calculated using the following relationship:

$$\mathrm{DQE} = \mathrm{SNR}_{\mathrm{out}}^2 / \mathrm{SNR}_{\mathrm{in}}^2$$

The DQE for a perfect digital detector is 1 or 100%. This means that there is no loss of information. Since $\mathrm{SNR}_{\mathrm{in}}$ takes into consideration the amount of exposure used and the $\mathrm{SNR}_{\mathrm{out}}$ considers the resultant image quality, the DQE indicates the detector performance in terms of output image quality and input radiation exposure used.

The DQE for CR is much better than film-screen (F-S) image receptors. This means that CR can convert an input exposure into a useful output image over a much wider range of exposures compared to F-S radiography. The DQE for "conventional CR detectors" is generally less than 30% [4]. Different manufacturers will report different DQE values for their respective CR systems. Interested readers should consult the CR specifications of the different manufacturers for their reported values.

3.10 Image Artifacts Overview

The notion of artifacts in F-S radiography is not new to technologists who can readily identify them, describe how they are produced, and even eliminate or reduce their impact on the image [9]. An *artifact* is "any false visual feature on a medical image that simulates tissue or obscures tissue" [9]. An earlier definition of an artifact is provided by Willis et al. who state that "an artifact is a feature in an image that masks or mimics a clinical feature" [10]. Artifacts therefore can be disturbing to radiologists and may even result in an inaccurate diagnosis. For CR, it is important that both radiologists and technologists be able to not only identify artifacts but also understand how they arise and how they can be reduced or removed from the image. In general, CR artifacts arise from the image acquisition process and

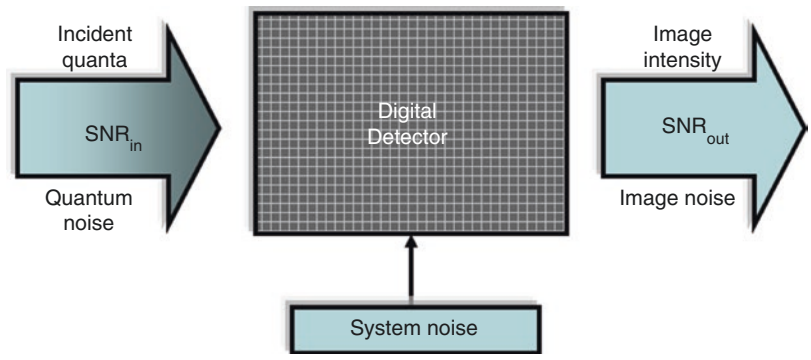

Fig. 3.14 The concept of the detective quantum efficiency (DQE). The digital takes an input exposure and converts it into a useful output image. The DQE is a measure of the efficiency with which the detector performs this task

include operator errors and the image processing system as well [9–13].

The types of artifacts arising from the image acquisition and image processing systems as described by Shetty et al. [13] are listed in Tables 3.2 and 3.3, respectively. The causes, appearance, and remedy are also listed in these two tables.

With respect to artifacts, the American Society of Radiologic Technologists (ASRT) suggests that "a best practice in digital radiography is to recognize image artifacts and prevent future artifacts from occurring by properly maintaining or acquiring service for the digital radiography equipment. In addition, a best practice in digital radiography is selection of the correct processing menu for an examination to ensure image quality" [14].

3.11 Continuous Quality Improvement Overview

The maintenance of equipment is an integral part of the daily activities in radiology since it is considered a component of a *continuous quality improvement* (CQI) program. The notion of CQI

Table 3.2 Image acquisition artifacts

Artifact	Causes	Appearance	Remedy
Twin artifacts	Two subsequent exposures on same imaging plate or double exposure	Duplication of images	Proper knowledge of usage of X-ray equipment
Uncollimated images	Improper collimation	Unsharp images	Proper collimation in accordance with cassette size and body part
Delayed scanning	Delay between acquisition and processing of image	Fading of image	Proper knowledge of radiographers to check that no delay occurs between acquisition and processing
Exposure through back of cassette	Poor basic knowledge of construction of cassettes	Various patterns of shading according to cassette design	Proper education of radiographers in handling of cassettes
Inappropriate exposure factors			
Overexposure	Improper exposure settings	Darkening of image	Proper exposure factors to be used based on body part and patient build
Underexposure	Improper exposure settings	Grainy image owing to quantum mottle	Knowledge of dynamic range and its limitations in computed radiography system
Improper grid usage	Usage of grids with low grid frequencies	Different types of moiré pattern	Usage of grids with 60 lines/cm or more; grid lines should run perpendicular to plate reader's laser scan lines
Scatter radiation	Cassette placed in vicinity of scattered radiation	Deterioration of quality of image and imprints of objects placed over cassette	Protect cassette from any unwanted radiation and erase cassettes before using
Care and carelessness	Mishandling of imaging plate during cleaning process	Kink marks	Cassettes and image plates should be handled with care
Light bulb effect	Backscattered radiation entering imaging plate from patient's bed due to increased exposure for obese patients or due to uncollimated X-ray	Darkening of lower and outer portions of an image relative to the remainder of image	Reduce backscatter by lowering the peak kilovoltage or by more precise collimation

Table 3.3 Image-processing artifacts

Artifact	Causes	Appearance	Remedy
Hardware-induced artifacts			
Imaging plate			
Artifacts due to cracks on imaging plats	Damaged imaging plates during frequent transportation	Cracks	Change imaging plate
Artifacts due to dust particles on imaging plate	Dust particles wedged over imaging plate	Focal radiopacities	Regular cleaning of imaging plates with ethyl alcohol
Roller artifacts			
Disparity artifact	Malfunctioning of rollers in digitizer	Defective scanning resulting in alteration in image contrast	Periodic cleaning of rollers in computed radiography reader
Damage of imaging plate due to rollers	Mechanical damaging of imaging plate during transport through rollers	Focal linear radiopacities	Warrants immediate cleaning of rollers
Dust over rollers	Dust deposited over imaging plate during transport through rollers	Multiple localized radiopacities	Maintenance and cleaning of rubber rollers by company service personnel twice yearly
Malfunctioning rollers	Slipping of feed rollers from transport assembly	Half-read image	Periodic cleaning and recalibration of feed rollers
Plate reader artifact	Dirt over light guide or beam deflector	Linear radiopaque line	Periodic cleaning of light guide by service personnel
Cassette-related artifact	Cracked or weakened lead coating on back of cassette	Linear radiolucent lines	Replacement of cassette
Software-induced artifacts			
Image transmission errors			
Communication error artifact	Power failure during image transmission	Missing lines or pixels in resulting image	Radiograph should be repeated
Data cable malfunctioning artifact	Failure of data cables in the power unit of digitizer or computed radiography reader	Alternating radiopaque and radiolucent lines	Replace data cables
Artifact due to improper erasure setting	Wrong body part selection or malfunctioning halogen bulbs	Residual image is left in imaging plate	Proper selection of body part for appropriate radiograph or halogen vapor bulbs must be changed

was developed by the Joint Commission on the Accreditation of Healthcare Organizations (JCAHO) in 1991. CQI ensures that every employee plays a role in ensuring a quality product. Prior to the introduction of the concept of CQI, other systems were in place to ensure quality patient care in hospitals. In radiology in particular, quality assurance (QA) and quality control (QC) programs are essential not only for optimizing the assessment and evaluation of patient care but also for monitoring the performance of equipment. QA and QC for digital radiography will be described further in Chap. 11.

3.11.1 Quality Assurance

Quality assurance (QA) is a term used to describe systems and procedures for assuring quality patient care. It deals specifically with quality assessment, continuing education, the usefulness of quality control procedures, and the assessment

of outcomes. QA deals with the administrative aspects of patient care and quality outcomes.

3.11.2 Quality Control

Quality control, on the other hand, is a component of QA and refers specifically to the monitoring of important variables that affect image quality and radiation dose. QC deals with the technical aspects (rather than administrative aspects) of equipment performance. The purpose of the procedures and techniques of CQI, QA, and QC is threefold: to ensure optimum image quality for the purpose of enhancing diagnosis, to reduce the radiation dose to both patients and personnel, and to reduce costs to the institution.

Quality control involves a number of activities that are of significance to the technologist, particularly if the technologist is in charge of the QC program. Such activities range from acceptance testing, routine performance, and error correction [9]. While *acceptance testing* is the first major step in a QC program and it ensures that the equipment meets the specifications set by the manufacturers, *routine performance* involves performing the actual QC test on the equipment with varying degrees of frequency (annually, semiannually, monthly, weekly, or daily). Finally *error correction* ensures that equipment not meeting the performance criteria or tolerance limit established for specific QC tests must be replaced or repaired to meet tolerance limits.

3.11.3 Parameters for QC Monitoring in CR

QC for CR has evolved from simple to more complex tests and test tools for use by radiology personnel to ensure that the CR equipment is working to ensure optimum image quality within the as low as reasonably achievable (ALARA) philosophy in radiation protection. It is not within the scope of this chapter to describe the elements of these QC programs; however, a few important points will be highlighted, and a more detailed description will be presented in Chap. 11.

The AAPM has recommended several testing procedures for CR QC (AAPM [15]), using specific tools developed for CR QC. A few of these testing procedures, for example, include physical inspection of the IP, dark noise and uniformity, exposure indicator calibration, laser beam function, spatial accuracy, erasure thoroughness, aliasing/grid response, positioning, and collimation errors, to mention only a few. Examples of these tools include screen-contact wire mesh pattern, anti-scatter grid, calibrated ion chamber, densitometer for hard copy film evaluation (in cases where film is used), manufacturer-approved cleaning solutions and cloths, two metric-calibrated 30-cm steel rulers, and low-contrast phantom to mention only a few.

3.11.4 Tolerance Limits or Acceptance Criteria

The AAPM has also established acceptance criteria for the recommended tests. They offer both qualitative and quantitative criteria. For example, they state that for CR IP uniformity response, the qualitative criteria should be a "uniform image without any visible artifacts with window/level adjustments." For details for the criteria and tests, the interested reader should refer to the AAPM Report No 93.

References

1. Lanca L, Silva A. Digital imaging systems for plain radiography. New York: Springer Science+Business Media; 2013.
2. Rowlands JA. The physics of computed radiography. Phys Med Biol. 2002;47:R123–66.
3. Neitzel U. Status and prospects of digital detector technology for CR and DR. Radiat Prot Dosim. 2005;14(1-3):32–8.
4. Seibert JA. Computed radiography technology. In: Goldman LW, Yester MV, editors. Specifications, performance, and quality assurance of radiographic and fluoroscopic systems in the digital era. College Park. Maryland: American Association of Physicists in Medicine (AAPM); 2004.
5. Fuji Photo Film Company. Fuji computed radiography technical review. No 14, Imaging Plate (IP). Tokyo, Japan. 2002; 1–23.

6. Flynn MJ. Processing digital radiographs of specific body parts. In: Samei E, editor. Advances in digital radiology, categorical course in diagnostic radiology physics. Illinois: RSNA; 2003.
7. International Electrotechnical Commission. IEC 62494-1 ed. 1 Medical electrical equipment: exposure index of digital X-ray imaging systems-part 1: definitions and requirements for general radiography. Geneva, Switzerland: IEC. 2008.
8. American Association of Physicists in Medicine (AAPM) Report No 116. An exposure indicator for digital radiography. College Park, MD: American Association of Physicists in Medicine; 2009.
9. Bushong S. Radiologic science for technologists. 11th ed. St Louis, MO: Elsevier; 2017.
10. Willis CE, Thompson SK, Shepard SJ. Artifacts and misadventures in digital radiography. Appl Radiol. 2004;33(1):11–20.
11. Cesar LJ, Schuelar BA, Zink FE, Daly TR, Taubel JP, Jorgenson LL. Artefacts in computed radiography. Brit J Radiol. 2001;4:195–20.
12. Honey ID, Mackenzie A. Artifacts found during quality assurance testing of computed radiography and digital radiography detectors. J Digit Imaging. 2009;22(4):383–92.
13. Shetty CM, Barthur A, Kambadakone A, Narayanan N, Kv R. Computed radiography image artifacts revisited. Am J Roentgenol. 2011;196:W37–47.
14. Herrmann TL, Fauber TL, Gill J, Hoffman C, Orth DK, Peterson PA, et al. Best practices in digital radiography. Albuquerque, NM: American Society of Radiologic Technologists (ASRT); 2012. p. 1–26.
15. American Association of Physicists in Medicine (AAPM). Acceptance testing and quality control of photostimulable storage phosphor imaging systems. Report No 93. 2006. College Park, MD.

Flat-Panel Digital Radiography

4

Abstract

The technical aspects of flat-panel digital radiography (FPDR) systems are described in detail in this chapter. The essential elements of these detectors include a definition of FPDR, system components, types of FPDR systems, design characteristics, operational principles, image processing, and imaging performance characteristics, as well as image artifacts and basic features of wireless digital detectors. The definition of these systems includes the use of amorphous silicon (a-Si) and amorphous selenium (a-Se) to produce digital images using a set of electrical components specifically designed to produce digital radiographic images in a manner different than computed radiography (CR) technology. Two common FPDR systems include the indirect and direct digital detector technologies. While the former produces images using an X-ray scintillator (X-ray conversion layer), followed by an amorphous silicon (a-Si) photodiode flat-panel layer, with a thin-film transistor (TFT) array for readout of the electrical charges by the photodiode array, the latter is based on the use of a photoconductor such as amorphous selenium (a-Se) to detect X-ray photons from the patient and convert them directly into electrical charges. A TFT array and associated electronics are also used to collect and store the changes for subsequent readout. The EI is also characteristic of FPDR systems and provide a visual cue as to the amount of exposure used for a particular examination and whether the exposure falls within guidelines suggested by the manufacturer. In addition pre-processing and post-processing present the raw data "for processing" and "for presentation," respectively. Finally, the chapter concludes with a description of imaging performance characteristics (such as spatial resolution, modulation transfer function, dynamic range, detective quantum efficiency, image lag, and artifacts) and wireless digital radiography detectors.

4.1 Introduction

In Chap. 3, the physics and technology of computed radiography (CR) imaging systems were described in detail. In review, a CR imaging system consists of at least four fundamental steps as shown in Fig. 4.1. First, the CR imaging plate (IP) is exposed to X-rays, after which it is placed into the CR image reader for signal extraction. Thirdly, a computer processes the signal and displays a digital image on a viewing monitor for diagnostic interpretation and possibly image post-processing.

The physical basis for the CR image formation is the IP detector that is made of a photostimulable phosphor (PSP) such as barium fluorohalide (BaFX-X is the halide which can be

E. Seeram, *Digital Radiography*, https://doi.org/10.1007/978-981-13-3244-9_4

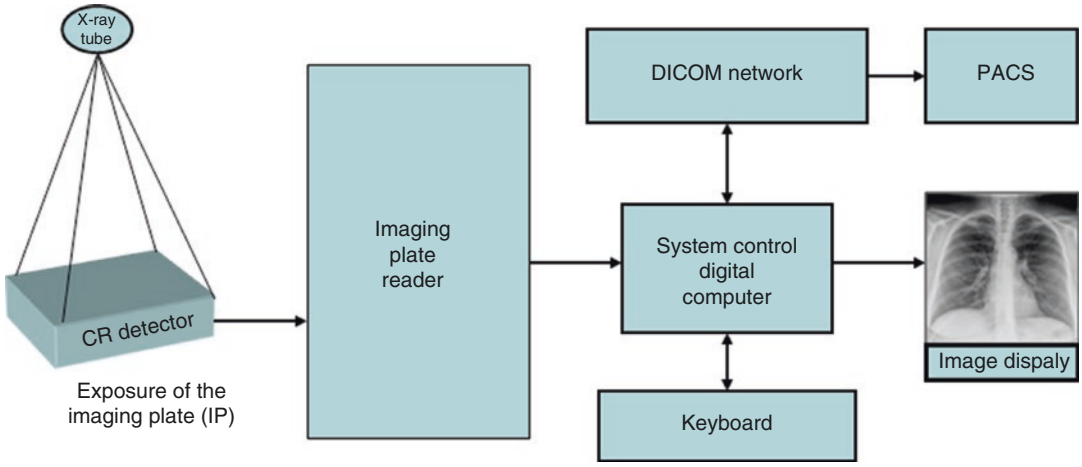

Fig. 4.1 The four fundamental steps in the CR process. After the IP is exposed to X-rays, it is taken to the image reader for latent image extraction. The IP is scanned by a laser beam, and the resulting signal is digitized and converted into an image. The image is displayed for viewing and interpretation and subsequently sent to PACS

a bromide, chloride, or iodide or a combination of these). After X-ray exposure of the IP PSP, a latent image is formed on the IP. To render this latent image visible, the IP is taken to the image reader that uses a laser beam to extract image information signal. This signal is subsequently digitized and sent to a computer for preprocessing; the computer-generated image is then displayed on a monitor for viewing and interpretation.

4.2 Limitations of CR

While CR has its advantages over film-screen (F-S) radiography, the most conspicuous being a wide dynamic range (Chap. 3), it also has several limitations. These include the following:

1. The X-ray detection efficiency of CR is inefficient, and this affects image quality and dose. The detection efficiency is also inferior to F-S radiography [1].
2. The spatial resolution of CR is less than F-S radiography. While the approximate spatial resolution for CR is about 6 line pairs/mm (lp/mm), it is about 8 lp/mm for F-S radiography [1].
3. CR IPs can easily be damaged. They can be dropped accidentally (during portable radiography or transported to the IP reader) and are susceptible to scratches and cracking. They can also be easily damaged when they are in the image reader.
4. CR IPs must be carried physically by the technologist to a separate image processor (image reader) for image data extraction.

The above limitations are some of the important considerations that provide the motivation for the development of other digital radiography systems. These systems are popularly referred to as digital radiography (DR) systems, using flat-panel detector technology. In this book the term *flat-panel digital radiography (FPDR)* will be used.

The purpose of this chapter is to describe the overall structure of flat-panel digital detectors and outline the physical principles and technology aspects of these detectors. In addition, several performance characteristics and other considerations will be highlighted.

4.3 What Is Flat-Panel Digital Radiography?

Flat-panel digital radiography detectors were introduced as early as 1995 for use in radiographic imaging [2]. While one system was based

on the use of amorphous silicon [2], the other was an amorphous selenium detector [3].

It is important to note here that other digital detectors for use in radiography were being used before 1995; however, they were not based on flat-panel detector technology. Two of these systems (CCD chip) were slot-scan digital detector which was introduced in 1990 and the selenium drum digital radiography system introduced in 1994 [4]. The selenium drum technology was developed as a dedicated system specifically for imaging the chest. The unit was called the Thoravision (Philips Medical Systems, The Netherlands). Selenium is a photoconductor, and it conducts electrons when struck by light or X-ray photons. Since there may still be a few systems currently in use, the fundamental principles will be reviewed briefly. Figure 4.2 shows how the selenium drum technology works. A positive change is first placed on the drum, which is then exposed to X-rays. The exposure changes the change distribution in the selenium to produce an electrostatic image stored on the drum. A probe then scans the electrostatic image to produce a voltage that is subsequently sent to an analog-to-digital converter (ADC), which digitizes the signal and sends the digital data stream to a digital computer for digital processing. The image is then displayed for viewing by the radiologist. The quality of the selenium drum image has been shown to be superior to that of both F-S radiography and CR imaging [4].

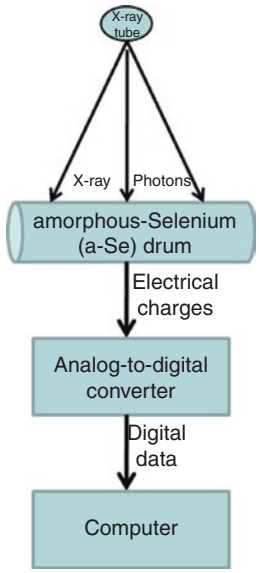

Fig. 4.2 A digital radiography imaging system that is based on selenium drum technology

4.3.1 Flat-Panel DR: System Components

A flat-panel DR imaging system is illustrated in Fig. 4.3. It is clear that there are several components that are coupled to the flat-panel detector. As shown in the diagram, the flat-panel digital detector is a single unit (a thin flat-panel device) that contains not only the flat-panel X-ray detection array but also the associated electronics as well. These include the preamplifiers, switching control, the central logic circuits, the ADCs, and internal memory. It is important to note that X-ray detection and digitization of the X-ray signal take place within the flat-panel detector. There is no need to take the flat-panel detector to a separate unit for signal readout as is a characteristic feature in CR. The different types of flat-panel detectors, imaging principles, and technology will be described in detail later in the chapter.

As can be seen in Fig. 4.3, the host computer acts as an interface between the flat-panel detector and the X-ray machine (X-ray tube, generator, and control panel) and other system components such as the image display device, network communication, and the image storage device. The host computer plays an important role in controlling X-ray production and signal readout from the flat-panel detector [5]. In addition, the host computer applies the appropriate image processing "to perform image correction and optimization for display" [6].

Other system components such as the image display, image storage, and network communication are responsible for several tasks. The computer output image is first displayed on a monitor for viewing, image post-processing, and interpretation by a radiologist. The image storage provides a repository for all images obtained from the flat-panel detector that are acceptable by the radiologist and are appropriately archived. Finally, the network component provides the

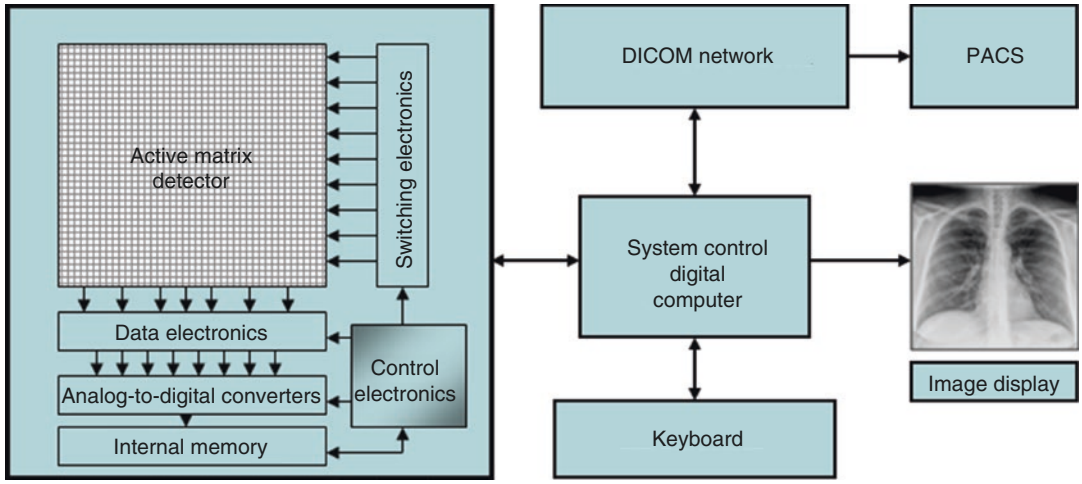

Fig. 4.3 A schematic representation of a flat-panel digital radiographic imaging system. The flat-panel detector contains all essential X-ray detection elements and electronics for image capture and retrieval, respectively

opportunity to distribute the images to interested parties located anywhere in the world.

4.4 Types of Flat-Panel Detectors

The use of flat-panel digital detectors in X-ray imaging is referred to as direct radiography [4]. Currently, there are two categories of flat-panel digital radiography detectors, namely, indirect detectors and direct detectors based on the type of X-ray absorber used (Fig. 4.4). While *indirect detectors* use a phosphor, *direct detectors* use a photoconductor, as is clearly illustrated in Fig. 4.4. With the above in mind, Yorkston [6] points out that "the rationale for this nomenclature is that the systems that use phosphors convert X-ray energy into electrical change through an intermediate stage of light photons, while those that use photoconductors convert the X-ray energy directly into electrical charge without the intermediate stage" [6]. In this respect therefore the terms indirect conversion and direct conversion have been used synonymously to refer to indirect detectors and direct detectors, respectively.

4.4.1 Indirect Digital Detectors: Technical Components

As illustrated in Fig. 4.4, there are two types of indirect digital detectors: the charge-coupled device (CCD) digital detector shown in Fig. 4.4a and the flat-panel thin-film transistor (TFT) digital detector (Fig. 4.4b). The most conspicuous difference between these two types of detectors is the technical component used to convert light into electrical signals.

4.4.1.1 CCD Digital Detectors

The CCD digital detector is based on an indirect conversion process and uses a CCD chip to convert light to electrical charge. The CCD digital detector is not classified as a flat-panel digital detector (the title of this chapter); however, because it is one system that is available commercially, it will be described here.

The main technical components of a CCD-based digital radiography detector are shown in Fig. 4.4a, which includes an X-ray absorber, light optics, and the CCD, which is a sensor (chip) for capturing the light. The CCD is also an electrical charge readout device and consists of several

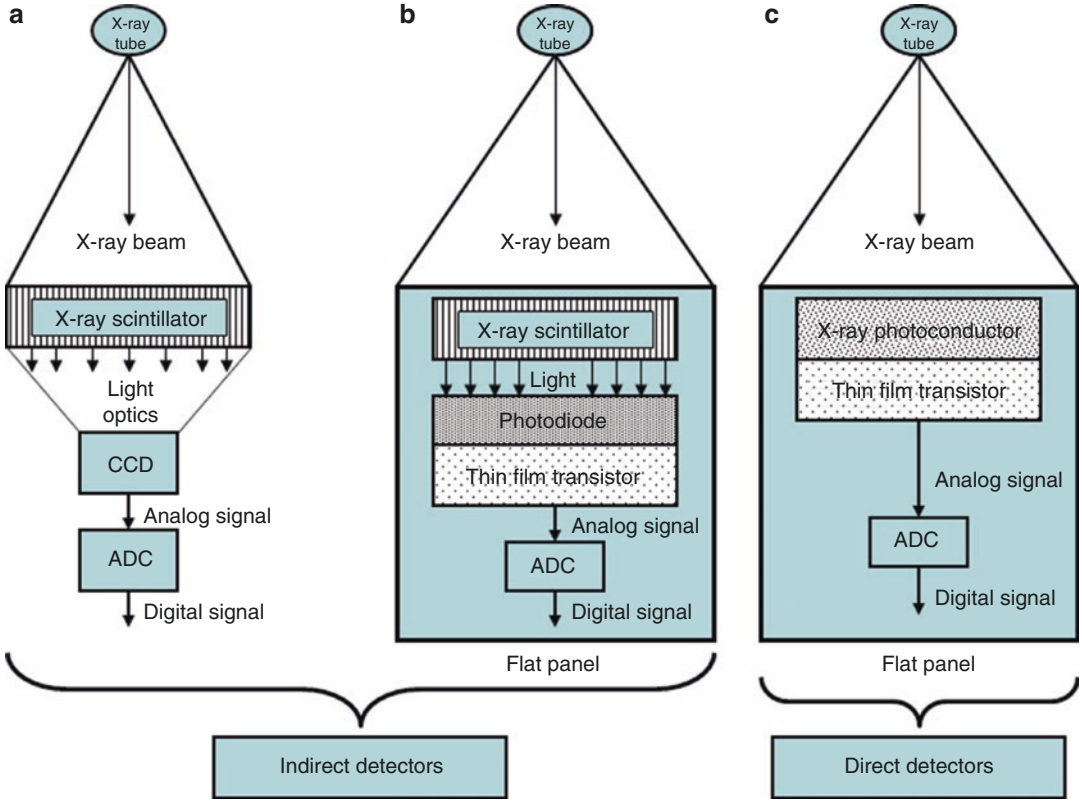

Fig. 4.4 Two categories of flat-panel digital radiography detectors, indirect detectors (**a** and **b**) and direct detectors (**c**). The CCD detector is not generally considered a flat-panel detector (see text for further explanation)

CCDs to increase the size of the detection area. Furthermore, there are three other components that are noteworthy: the scintillation screen (detects X-rays and converts them into light), the light collection optics, and, finally, an array of CCDs, also referred to as the CCD camera. Some CCD detectors can consist of a fiberoptic-coupled CCD system, while others may use a lens-coupled CCD or a fiberoptic-coupled scanning array.

4.4.1.2 Indirect Flat-Panel TFT Digital Detectors

An *indirect flat-panel* digital detector is based on an indirect conversion process and uses several physical components to convert X-rays first, into light that is subsequently converted into electrical charges. These components include an *X-ray scintillator* (X-ray conversion layer), followed by an *amorphous silicon (a-Si) photodiode* flat-panel layer, with a *thin-film transistor (TFT)* array for readout of the electrical charges by the photodiode array. The organization of these components is illustrated in Fig. 4.5.

The X-ray scintillator layer used in the indirect flat-panel digital detector is usually *cesium iodide* (CsI) or gadolinium oxysulfide (Gd_2O_2S). These phosphors are not new to X-ray imaging for they have been used in X-ray image intensifiers (CsI) and in rare-earth intensifying screens (Gd_2O_2S). An interesting design fabrication of these phosphors is the manner they are deposited into the a-Si photodiode array. While CsI crystals are deposited in a needle-like fashion (structured

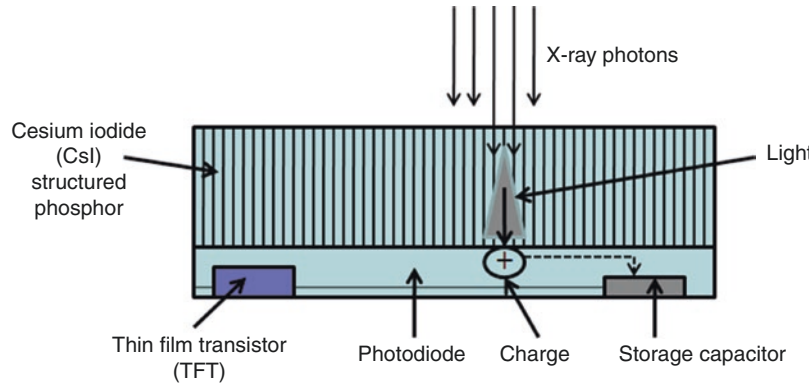

Fig. 4.5 The main system components of an amorphous-Si (a-Si) photodiode TFT indirect flat-panel digital detector

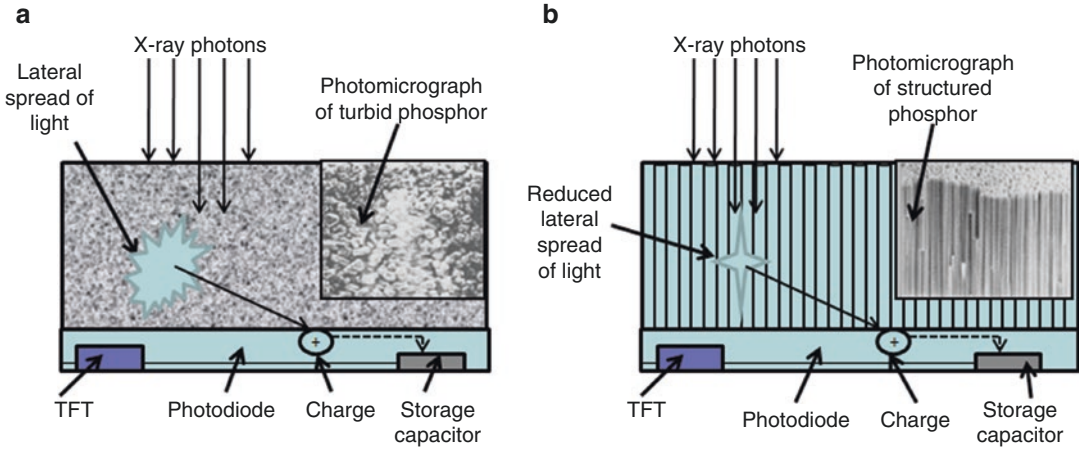

Fig. 4.6 A schematic of the function of a turbid (powered) phosphor (**a**) and a structured phosphor (**b**) indirect flat-panel digital detectors. While the turbid phosphor design results in a lateral spread of light, the structured phosphor "needle-like" design reduces the lateral spread of light which serves to improve the spatial resolution of the image

phosphor) and run in the direction of the X-ray beam, Gd_2O_2S crystals are deposited as powdered particles (turbid phosphor). While the former is sometimes referred to as a structured scintillator, the latter is referred to as an unstructured scintillator. In addition, powdered phosphors produce lateral spreading of light, which destroys the spatial resolution of the image. Structured phosphors like the CsI needles on the other hand reduce the lateral dispersion of light, thus improving the spatial resolution of the image. This effect is clearly illustrated in Fig. 4.6.

Following the X-ray scintillator (X-ray detection medium) is an a-Si photodiode flat-panel array. This component is made by a process referred to as "plasma-enhanced chemical vapor deposition" where "layers of a-Si are deposited onto a thin glass substrate (typically ~0.7 mm) …" [6]. The purpose of the a-Si photodiode layer is to convert the light from the X-ray detection scintillator into electrical charges. Adjacent to the a-Si photodiode layer is a thin-film transistor (TFT) array, as well as storage capacitors and associated electronics (Fig. 4.6). The purpose of the capacitor is to collect and store the electrical charge produced in the a-Si photodiode array. In this regard the flat-panel described above is also referred to as an indirect conversion TFT digital detector. The flat-panel design will be described later in the chapter.

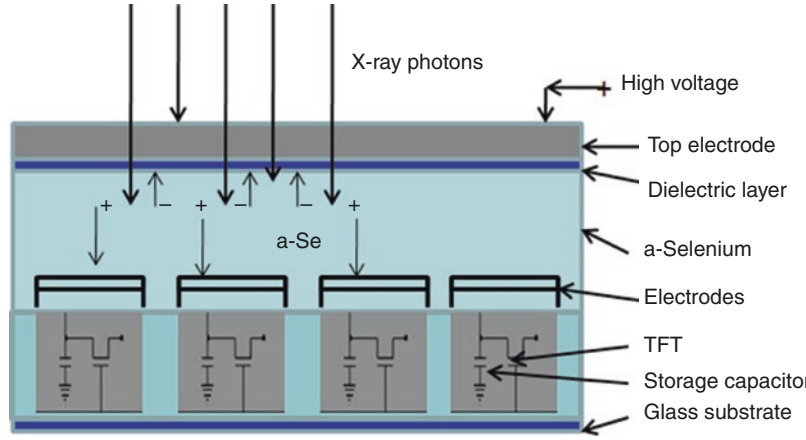

Fig. 4.7 The main system components of an amorphous-Se (a-Se) TFT direct flat-panel digital detector

4.4.2 Direct Digital Detectors: Technical Components

A cross-sectional diagram of a *direct flat-panel TFT digital detector* is shown in Fig. 4.7. This detector consists of several components such as a source of high voltage, top electrode, dielectric layer, photoconductor, collection electrode, TFT, storage capacitor, and the glass substrate. It is not within the scope of this book to describe the details of each of these components; however, the basics of the photoconductor will be highlighted.

The photoconductor is *amorphous selenium (a-Se)* although other photoconductors such as lead oxide, lead iodide, thallium bromide, and gadolinium compounds can be used [4]. The use of a-Se is now popular since it offers excellent X-ray photon detection properties and provides images with very high spatial resolution. The photoconductor detects X-ray photons from the patient and converts them directly into electrical charges. The TFT and associated electronics (e.g., capacitors) collect and store the changes for subsequent readout.

The operational principles of indirect and direct flat-panel TFT digital detectors will be described later in this chapter. The next section of this chapter deals with the design (or construction framework) of the flat-panel TFT digital detector. The photoconductor detects X-ray photons from the patient and converts them directly into electrical charges. The TFT and associated electronics (capacitors for example) collect and store the charges for subsequent readout.

The operational principles of indirect and direct flat-panel TFT digital detectors will be described later in this chapter. The next section of this chapter deals with the design (or construction framework) of flat-panel TFT digital detector.

4.5 Design Characteristics of Flat-Panel Detectors

4.5.1 Configuration of the Flat-Panel

The flat-panel TFT digital detector is designed as a *matrix of detector elements*, each of which can be regarded as a pixel and constructed as shown in Fig. 4.8. This design principle is referred to as large area integrated circuit. The matrix, also referred to as an *active matrix array*, consists of rows and columns which play a role in addressing and readout of the signal from each pixel. As seen in Fig. 4.8, each pixel contains a TFT (switch), a storage capacitor, and a sensing area, referred to as the sensing/storage element. The sensing area will detect the light from the CsI scintillator in the indirect flat-panel TFT detector or, in the case of a direct flat-panel TFT detector, X-ray photons passing through the patient.

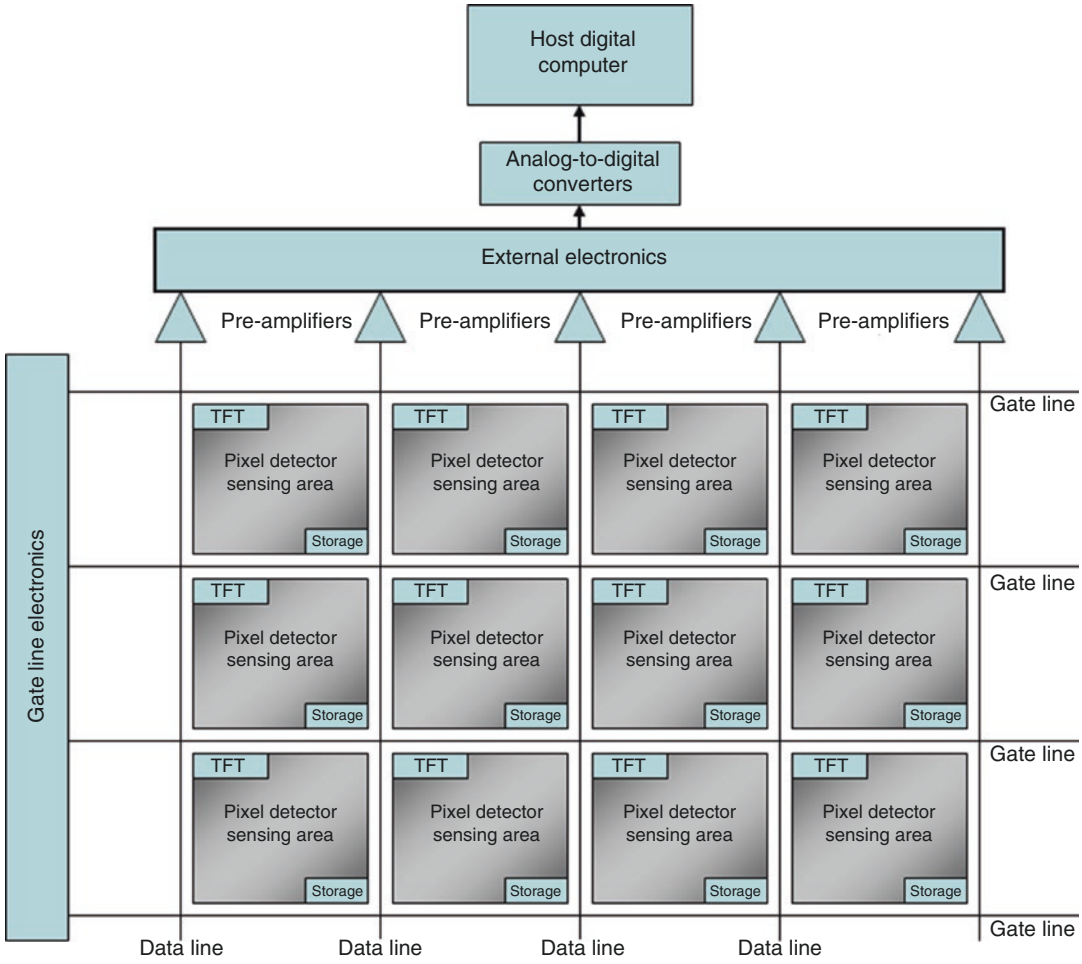

Fig. 4.8 The basic construction of a flat-panel digital detector showing the layout of the major system components. The flat-panel detector captures radiation from the patient and converts it into an electrical signal which is subsequently digitized by the analog-to-digital converters. The digital data are sent to the host computer for image processing (see text for further explanation)

In particular, the sensing/storage element of an indirect flat-panel TFT detector is the photodiode (photosensitive storage element), while it is the storage capacitor (capacitive storage element) in the direct flat-panel TFT detector that uses the a-Se photoconductor [5].

In addition to the matrix of pixels, there are other electronic components that are included in the flat-panel detector (gateline or data line). These include switching electronics to activate each row of pixels and electronic amplifiers and associated electronic devices (multiplexer) for signal readout from each column of pixels. The electronic signal (analog signal) is subsequently sent to the *analog-to-digital converter (ADC) for* digitization. The ADC is also included in the flat-panel detector. Finally, the digital data stream is sent to a digital computer for digital image processing.

4.5.2 Dimensions of the Detector and Components

As noted earlier, the flat-panel TFT digital detector consists of a matrix of pixels

(individual elements). Since there are different sizes of body parts to be imaged, different sizes of detectors are commercially available for clinical imaging. Typical detector dimensions, for example, include 43 × 43 cm, 30 × 40 cm, and 18 × 18 cm.

The matrix size also varies depending on the size of the detector. Typical matrix sizes include 1920 × 1536, 2000 × 2500, 2736 × 2736, 2560 × 3072, 2688 × 2688, 2836 × 2336, 3121 × 3121, etc.

The pixel size and spacing (*pixel pitch* = the distance from the midpoint of one pixel to the midpoint of the adjacent pixel) determine the spatial resolution of the image. The number of pixels can be obtained by multiplying the dimensions of the matrix size. For example, the number of pixels in a 2688 × 2688 matrix is 7,225,344 pixels. In general, pixel sizes can be 139 μm, 143 μm, 160 μm, 162 μm, 167 μm, and 200 μm.

4.5.3 The Fill Factor of the Pixel

An important feature of the pixel in the flat-panel TFT digital detector active matrix array is the *fill factor* as shown in Fig. 4.9. A pixel contains generally three components; the TFT, the capacitor, and the sensing area. The sensing area of the pixel receives the data from the layer above it that captures X-rays that are converted to light (indirect flat-panel detectors) or electrical charges (direct flat-panel detectors). The fill factor then is defined as the ratio of sensing area of the pixel to the area of the pixel itself [1] and can be expressed as:

$$\text{Fill factor} = \frac{\text{Sensing area of the pixel}}{\text{Area of the pixel}}$$

The fill factor is also expressed as a percentage, where a fill factor of 80% means that 20% of the pixel area is occupied by the detector electronics with 80% representing the sensing area.

The fill factor affects both the spatial resolution (detail) and contrast resolution (signal-to-noise ratio) characteristics of the detector [1]. Detectors with high fill factors (large sensing areas) will provide better spatial and contrast resolution than detectors with low fill factors (small sensing area).

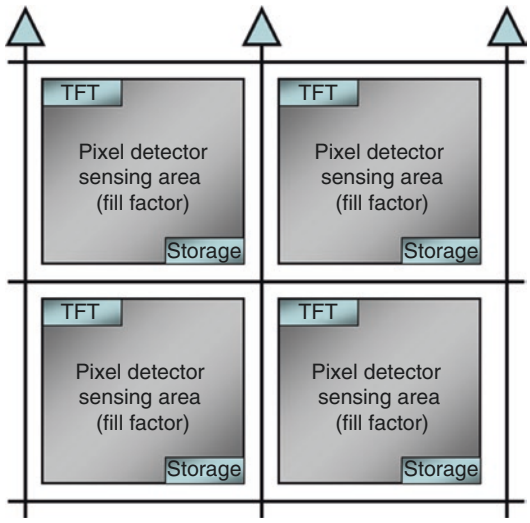

Fig. 4.9 The sensing area of a single pixel of a flat-panel digital detector is referred to as the fill factor (see text for further explanation)

4.6 Principles of Operation

The principles of operation of digital detectors based on CCD and flat-panel technologies are complex and demand a good understanding of the physics of scintillation phosphors, photoconductors, and semiconductors, as well as electronics. Since this is not a physics or electronics textbook, it is not within the scope here to describe the physics or electronics in any detail. In addition, some of the technical aspects of these detectors are propriety and are not disclosed to the public; therefore only a generalized description of how these detectors work will be presented.

The topics described in this section will focus on how the CCD detectors and flat-panel detectors work (from exposure of the detector to readout of the electrical charges that create a latent image), exposure latitude of flat-panel digital detectors, exposure indicators, and image processing.

4.6.1 CCD Digital Detectors

There are various design structures of the CCD digital detector. These are indirect conversion detectors using the CCD to collect light from the scintillator as illustrated in Fig. 4.4a. The CCD silicon chip is made up of millions of discrete pixels forming a matrix array of pixels. Light from the scintillator falls upon each pixel to produce electron hole pairs (charges) in direct proportion to the amount of light falling upon it. During readout, the system electronics provide a systematic collection of changes on each chip in a manner referred to as a "bucket brigade." The charge from each pixel in a row is transferred to the next row and subsequently down all the columns to the final readout row. The charge pattern from the readout row of pixels is sent to the ADC for digitization. The digital data stream is sent to a digital computer for processing to produce a digital image.

One of the limitations of the CCD digital detector is related to the light optics (lenses, mirrors, and fiber optics shown in Fig. 4.4a) used to couple the X-ray scintillator to the CCD silicon chip. Specifically, the optical system reduces the output image from the scintillator phosphor to the size of the CCD array. This minification or demagnification reduces the image quality such as a reduced signal-to-noise ratio [4]. In this respect, efforts have been made to design other CCD detectors to deal with image quality degradation with CCD detectors that have a small area. More studies are needed to assess the objective imaging performance of these detectors.

4.6.2 Flat-Panel TFT Digital Detectors

As described earlier in the chapter, there are two types of flat-panel TFT digital detectors for use in radiographic imaging. These are the indirect flat-panel TFT detectors (Fig. 4.4b) and the direct flat-panel TFT detector (Fig. 4.4c). While indirect detectors use a light-sensitive photodiode (to capture light from the scintillator phosphor such as CsI) to produce electrical charges, the direct detectors use an amorphous-Se (a-Se) photoconductor to convert X-rays directly into electrical charges.

The physics of the interaction of X-rays with scintillators, photodiodes, and photoconductors will not be described in this book. Such description can be found in any good radiologic physics textbook, such as Bushberg et al. [7]. Furthermore, a radiologic physics course for technologists should address the basic physics of how X-rays interact with these materials.

One important physical concept that is essential to understanding the rationale for the use of specific X-ray scintillators and a-Se in digital detectors is that of X-ray attenuation or absorption. Recall from physics that X-ray attenuation depends on several factors such as the atomic number (Z), density, and thickness of the attenuating materials, for example. As noted earlier, X-ray scintillators are CsI and Gd_2O_2S for indirect flat-panel TFT detectors and a-Se for direct flat-panel TFT detectors.

A conspicuous feature of these detectors is that they are commonly used in radiography, fluoroscopy, and angiography that are based on the use of the Brems or continuous X-ray spectrum. This spectrum ensures that a wide range of voltages (say from 50 kVp to 140 kVp) can be used for the varying types of body parts imaged in three modalities mentioned above. In addition, flat-panel detectors are also used in digital mammography (Chap. 8) in which a dedicated X-ray tube is used to produce a specific spectrum (discrete or characteristic spectrum) in which the range of energies is optimized to image the breast. Recall that mammographic kV can range from 26 to32 kV [7].

Additionally the Z for CsI is about 54 (Z for Cs = 55; Z for I = 53), and the thickness of the CsI scintillator layer is about 600 μm. These characteristics make this type of detector suitable for radiography, fluoroscopy, and angiography. For mammographic imaging, a thinner layer of CsI, about 150 μm, combined with the mammographic X-ray spectrum, are essential requirements for optimum imaging. For a-Se digital detectors, apart from the k-edge characteristic that plays an important role in X-ray absorption

(see any radiographic physics textbook), a thickness of about 250 μm is required for mammography, while about 1000 μm thickness is essential for radiographic and angiographic imaging [1].

Two processes, the detection and conversion of X-rays into electrical charges and subsequent readout of the charges, characterize the operating principles of flat-panel digital detectors. Each of these two processes will now be described.

Indirect flat-panel TFT digital detectors utilize a scintillator layer such as CsI to first convert X-ray photons into light photons. Secondly, these light photons strike the a-Si photodiode layer, which converts them into electrical charges. On the other hand, for direct flat-panel digital detectors, X-ray photons fall upon the a-Se photoconductor layered on top of a matrix of a-Si TFT array. In this detector, an electric field is created between the top electrode and the TFT elements. As X-rays strike the a-Se, electrical charges are created, and the electric field causes them to move toward the TFT elements where they are collected and stored. In both of these two processes, the distribution of the charges in the matrix array of pixels represents the so-called latent image. It is at this point that all charges must be read out. This is accomplished by what is referred to as the readout electronics.

The flat-panel TFT digital detector uses complex and sophisticated electronic circuitry to read out the electrical charges produced and stored in the matrix array. This is a very systematic process also referred to as an active matrix readout where pixels are read out row by row [5].

Before the flat-panel detector can be used for an X-ray examination, it must be prepared prior to the X-ray exposure. This preparation is referred to as initialization. The interested reader may refer to the description provided by Yorkston [6].

For the flat-panel TFT digital detectors used in fluoroscopy, the readout process and readout electronics are somewhat different than the detectors used for radiographic applications. While radiography produces static images, fluoroscopy produces dynamic images to demonstrate motion. Digital fluoroscopy will be described in Chap. 7.

4.6.3 Exposure Latitude

The *exposure latitude* in radiographic imaging is a concept that examines the response of the image receptor (screen-film detector or digital detector) to the radiation falling upon it. Exposure latitude for computed radiography detectors (photostimulable phosphor imaging plate) was described in Chap. 3. Essentially, the exposure latitude for digital detectors (CR and DR detectors) is much wider (0.1–1000 μGy) compared to the screen-film receptors. The major advantage of such wide dynamic range is that the detector can respond to different levels of exposure (low to high) and still provide an image that appears acceptable to the observer, as is clearly illustrated in Fig. 4.10. On the other hand, a significant disadvantage of this digital detector response is that while higher doses provide excellent image quality, the patient dose increases. Additionally, since operators and observers always tend to favor excellent image quality, operators may often use a higher exposure than is normally required for a particular examination. This situation is referred to as "exposure creep" or simply "dose creep" [8, 9].

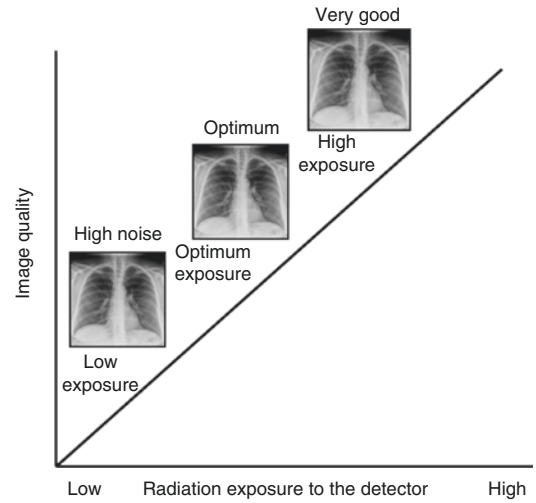

Fig. 4.10 The major advantage of such wide dynamic range is that the detector can respond to different levels of exposure (low to high) and still provide an image that appears acceptable to the observer

4.6.4 Exposure Indicator

The *exposure indicator* (EI) or exposure index (EI) as it is sometimes referred to was described in Chap. 3 and repeated here for emphasis since it is a useful tool to address the problem of "exposure creep." The exposure indicators for CR systems were explained in Chap. 3. In review, the EI is a numerical value usually displayed on the digital image to indicate the exposure to the digital detector. Therefore the EI provides a visual cue as to the amount of exposure used for a particular examination and whether the exposure falls within guidelines suggested by the manufacturer. For example, while the old FPD EIs of Philips Healthcare for 5 μGy, 10 μGy, and 20 μGy are 200, 100, and 50, those for the Siemens FPD at 5 μGy, 10 μGy, and 20 μGy are 500, 1000, and 2000. There is a "lack of standardization" of EIs for these two systems (and other systems as described in Chap. 3), and this can create problems on the intelligent use of EIs in a clinical environment [9]. The standardized EI will be described in detail in Chap. 5.

4.7 Image Processing: Optimizing the Display of the Image

The fundamental principles of digital image processing were described in Chap. 2. Subsequently, specific image processing techniques used in CR imaging systems were discussed in Chap. 3. Essentially, the image processing algorithms used in CR are used in flat-panel digital imaging systems.

The purpose of image processing in systems using flat-panel digital detectors is twofold:

1. To provide corrections to the raw digital data obtained from the detector. This will reduce artifacts in image post-processing and will be described subsequently.
2. To optimize the display of the image that is presented to the radiologist for diagnostic interpretation. This means that contrast and sharpness are enhanced, while noise can be reduced.

4.7.1 Image Processing Stages

There are two image processing stages that are used to accomplish the two tasks listed above: the image *pre-processing* stage and the image *post-processing* stage, as illustrated in Fig. 4.11. The raw digital data from the detector has low image contrast simply because of the wide exposure latitude of the digital detector. Therefore the raw digital data is first subject to pre-processing and presented as a DICOM *"for processing" image*. This image then undergoes digital image post-processing, and the resultant post-processed image is labeled a *"for presentation" image* [8]. While pre-processing operations deal with applying corrections to the raw data, post-processing operations address the appearance of the image displayed on a monitor for viewing and interpretation by a radiologist.

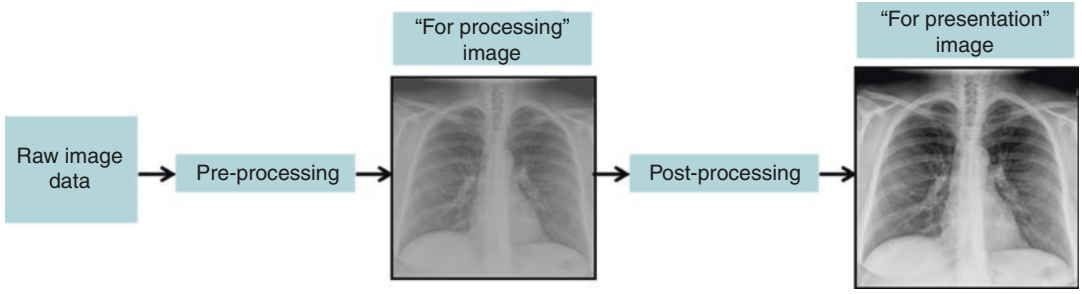

Fig. 4.11 The framework for processing the raw digital data for image display and viewing by an observer (Courtesy of Kerry Krugh, PhD, Medical Physicist, University of Toledo Medical Center, Toledo, Ohio)

Pre-processing techniques are intended to correct the raw data collected from bad detector elements that would create problems in the proper functioning of the detector. The image obtained initially from the detector referred to as a flat-field image may contain artifacts due to the bad detector elements. Examples of these artifacts are shown in Fig. 4.12a and include those due to dead or bad pixels, bad column of pixels, and so forth. These artifacts can be corrected by a pre-processing technique referred to as *flat-fielding* [8] as clearly seen in Fig. 4.12b. This correction process is popularly referred to as system calibration, and it is an essential requirement to ensure detector performance integrity. The chest image shown in Fig. 4.13 demonstrates the column artifact arising from a bad column of pixels and provides a rationale for ongoing calibration of the digital imaging system.

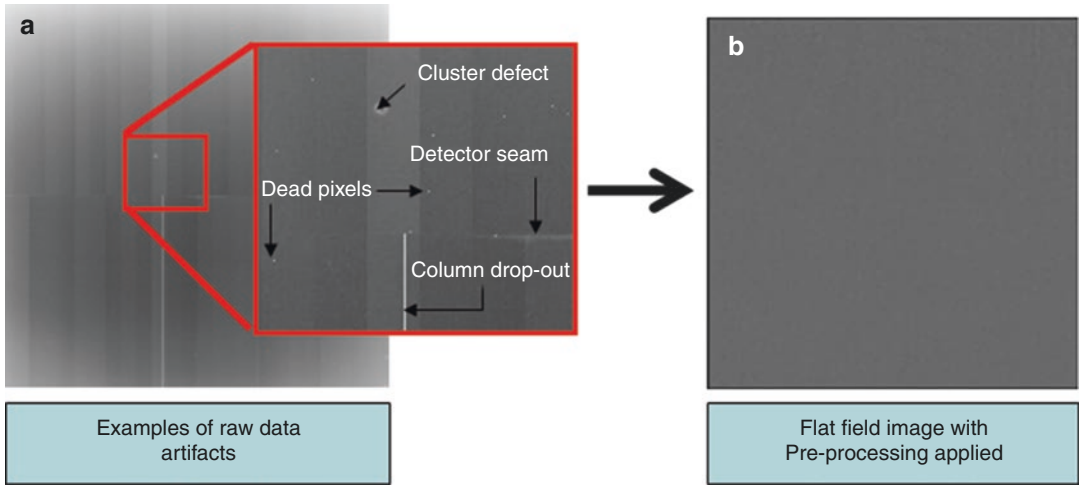

Fig. 4.12 Examples of image artifacts arising from bad detector elements (**a**) and flat-field image after corrections applied (**b**) (Courtesy of Kerry Krugh, PhD, Medical Physicist, University of Toledo Medical Center, Toledo, Ohio)

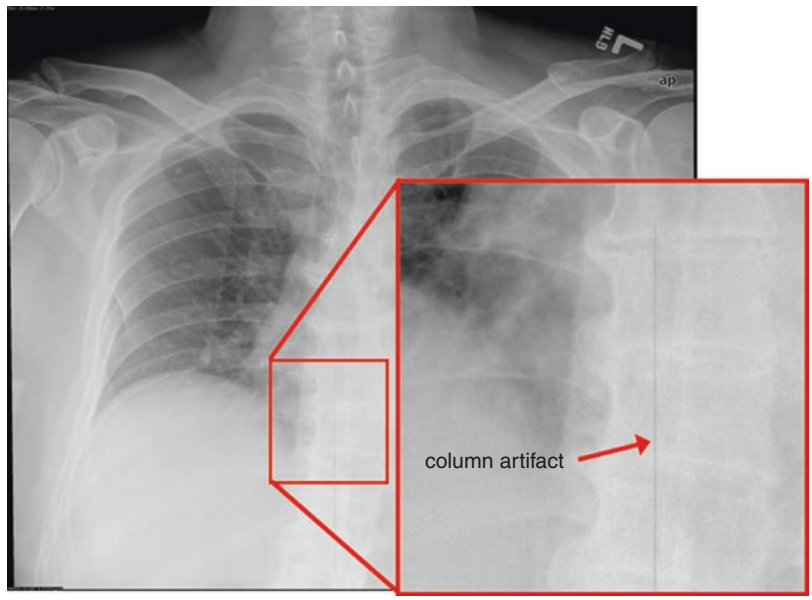

Fig. 4.13 The column artifact due to bad column of pixels on the flat-panel detector (Courtesy of Kerry Krugh, PhD, Medical Physicist, University of Toledo Medical Center, Toledo, Ohio)

The purpose of post-processing is clearly shown in Fig. 4.14, where the "for processing" image is converted into the "for presentation" image that has better contrast. The poor contrast of the "for processing" image in Fig. 4.14a is due to the wide exposure latitude of the digital detector. This image is subsequently subject to image post-processing and presented with improved contrast (Fig. 4.14b) to suit the viewing needs of the radiologist. As shown in Fig. 4.15, the wide exposure latitude (linear response) is converted into the well-known characteristic curve (nonlinear response) to produce an image with improved contrast (contrast enhancement). This is the "for presentation" image.

There are several steps in image processing to get from the "for processing" image to the "for presentation" image. These steps were described in Chap. 2 and then again in Chap. 3 for CR systems. In review, the steps include "exposure

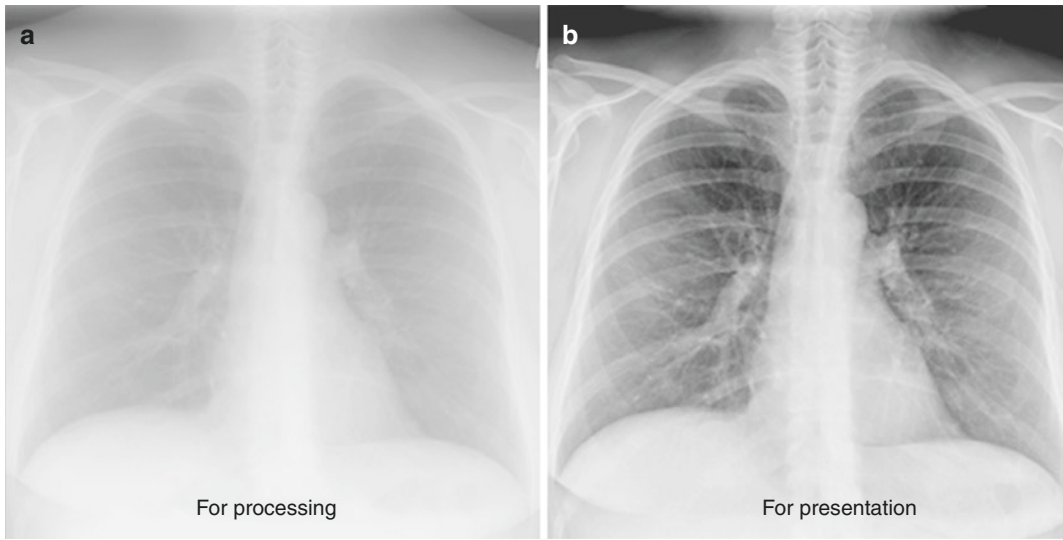

Fig. 4.14 The visual effect on image quality after post-processing applied to the "for processing" image (Courtesy of Kerry Krugh, PhD, Medical Physicist, University of Toledo Medical Center, Toledo, Ohio)

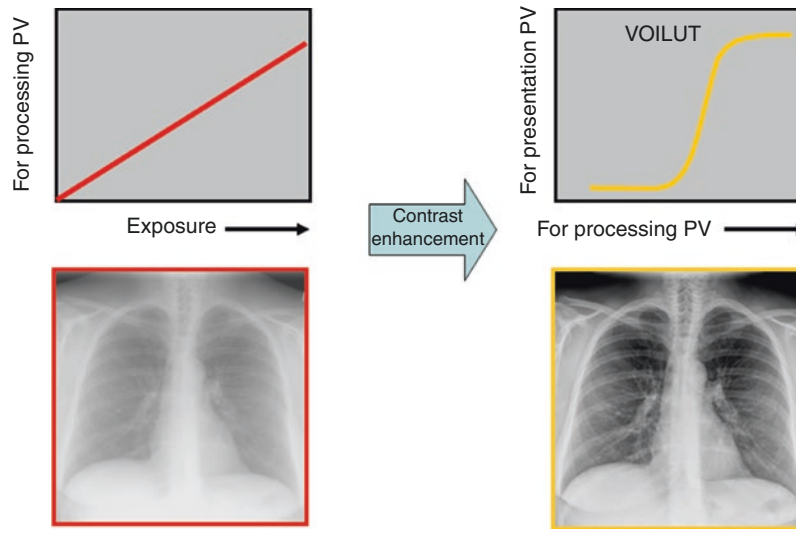

Fig. 4.15 The fundamental process of producing an image of optimum image contrast (Courtesy of Kerry Krugh, PhD, Medical Physicist, University of Toledo Medical Center, Toledo, Ohio)

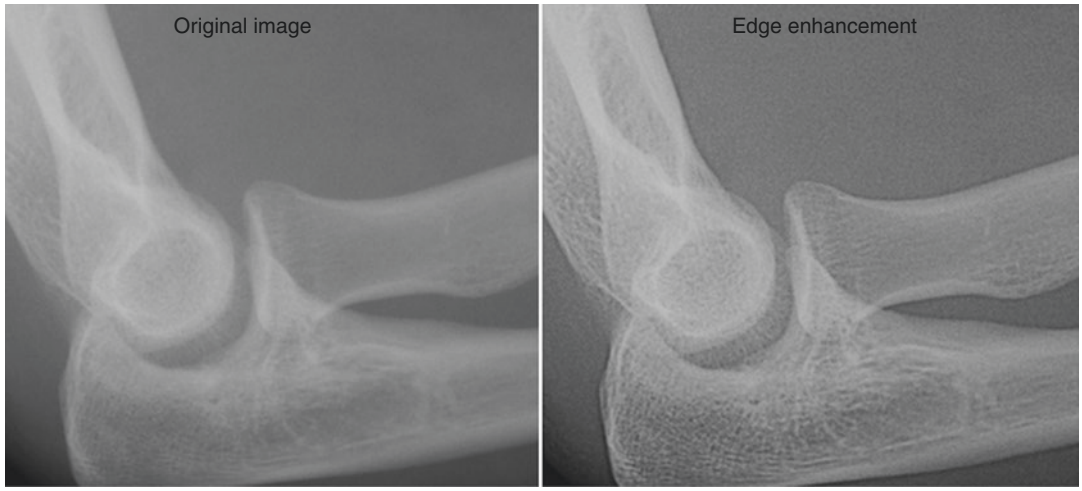

Fig. 4.16 The use of an edge enhancement image processing algorithm to sharpen the original image (Courtesy of Kerry Krugh, PhD, Medical Physicist, University of Toledo Medical Center, Toledo, Ohio)

recognition and segmentation of the pertinent pre-processed image data. These two steps find the image data by using image processing algorithms and *histogram* analysis to identify the minimal and maximal useful values according to the histogram-specific shape generated by the anatomy. This usually requires the user to choose the correct examination-specific processing algorithm" [8].

The next step involves *scaling the histogram* based on the exposure falling on the detector. If the exposure is too low (underexposure), for example, the raw data is adjusted to generate an image that looks pleasing to the observer.

The third step in image processing is *contrast enhancement* where the adjusted or scaled raw data values are mapped to the "for presentation" values to display an image with optimum contrast and brightness (Fig. 4.15). The displayed image can be post-processed further to enhance image detail or sharpness and to reduce image noise, using the spatial frequencies (high and low frequencies contained in the image). While the high frequencies contain fine details of the image, low frequencies contain the contrast information in the image. By using a high-pass filter, low frequencies are suppressed, and the image details are enhanced, and the image appears sharper. This is an *edge enhancement* algorithm and its effect on the image is shown in Fig. 4.16. Too much edge enhancement, however, produces image noise and creates the "halo" effect as shown in Fig. 4.17.

In a similar manner, a low-pass filter can be used to suppress the high frequencies to smooth (blur) the image. Such smoothing operations can help the radiologist to make a diagnosis, since they serve to reduce the noise present in the image. These algorithms were described in Chaps. 2 and 3.

4.7.2 Image Display Optimization

Just as exposure technique factors (mAs and kV) should be optimized to meet the requirements of the ALARA (as low as reasonably achievable) philosophy, image display should be optimized so that the best possible image is available to the radiologist for diagnostic interpretation.

There are several factors that affect the display of an image on the monitor, and it is not within the scope of this book to describe these in any detail. The interested student should refer to Bushong [1] for a comprehensive coverage of image display and viewing the digital image.

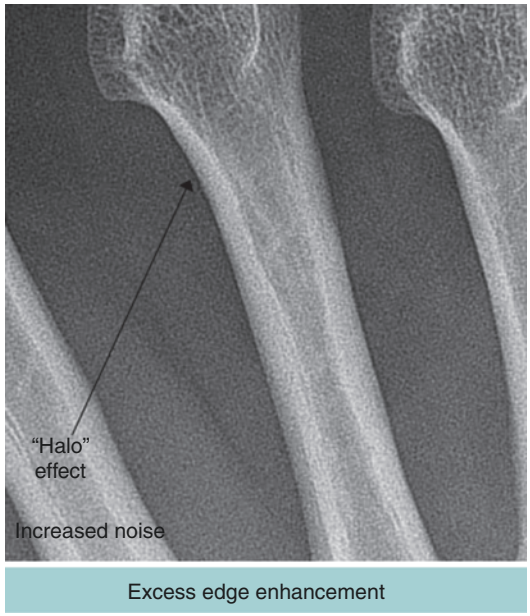

Fig. 4.17 Excess edge enhancement can result in an artifact referred to as the "halo effect" (Courtesy of Kerry Krugh, PhD, Medical Physicist, University of Toledo Medical Center, Toledo, Ohio)

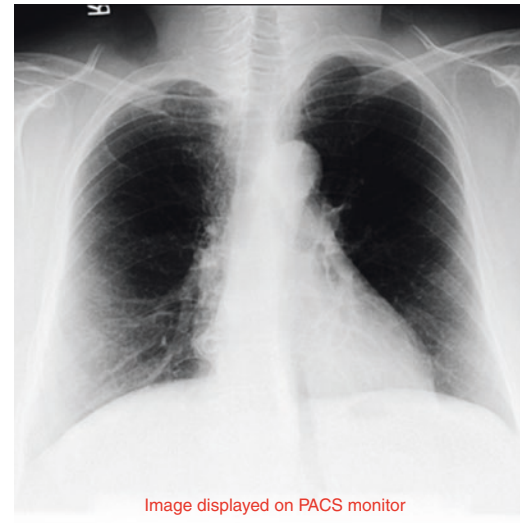

Fig. 4.18 The effect on the image at the PACS workstation monitor if the technologist workstation monitor is not calibrated to the DICOM GSDF. The image appears to have lost its optimum contrast (Courtesy of Kerry Krugh, PhD, Medical Physicist, University of Toledo Medical Center, Toledo, Ohio)

It is mandatory that technologists recognize that the effective use of image post-processing is a fairly complex process and that image processing is not intended to correct the poor images routinely due to errors that may have been made during image acquisition. If a technologist has to post-process every single image to improve its quality, this may be a sign that indicates a problem with the imaging system. Furthermore, different anatomical regions should be processed with their respective algorithms. For example, a chest image should not be processed with a shoulder algorithm, even though this may have a positive impact on the image displayed on the technologist workstation.

Problems arising as a result of image post-processing may also relate to the differences in the acquisition workstation and the higher-quality PACS workstation used by the radiologist. These workstations may have different display resolution and luminance characteristics, for example. In addition, the acquisition workstation may not be calibrated to a standard called the *DICOM Grayscale Standard Display Function* (GSDF), whereas the PACS radiologist workstation is calibrated. The result of this is shown in Fig. 4.18, where an image post-processed on the technologist acquisition monitor (not calibrated to GSDF) appears much differently (higher contrast) on the PACS workstation monitor. Optimization of the image displayed on monitors for viewing by both technologists and radiologists demands a good working knowledge of not only image processing but also display workstation (acquisition and PACS) principles and technology.

4.8 Imaging Performance Characteristics

The imaging performance of a detector generally refers to the "ability of the detector to produce a high-quality X-ray image" [5]. There are several characteristics affecting image quality. These include spatial resolution (sharpness), modulation transfer function (MTF), dynamic range, detective quantum efficiency (DQE), image lag, ghosting, and artifacts created by a host of factors, such as dead pixels. A complete description of these characteristics is beyond the scope of

this chapter; however, it is noteworthy to describe the essential elements of the "most important" ones. These include spatial resolution, dynamic range, MTF, DQE, and image lag. Each of these will be reviewed briefly, without resorting to any underlying physics and mathematics.

4.8.1 Spatial Resolution

Spatial resolution is the ability of the imaging system to resolve fine details present in an object. It also refers to the sharpness of the image. For digital imaging systems, the spatial resolution depends on the size of the pixels in the matrix. Smaller pixels will produce images with better spatial resolution compared with larger pixels.

Measuring the spatial resolution is a complicated process that involves at least three methods. These include imaging a bar test pattern, a sharp-edged object, or a narrow slit. While the image of the bar test pattern is easy to interpret visually for the sharpness of the lines in the bar test pattern (Fig. 4.19), the latter two are more complicated.

According to Dr. John Yorkston Ph.D, Senior Research Scientist, Clinical Applications

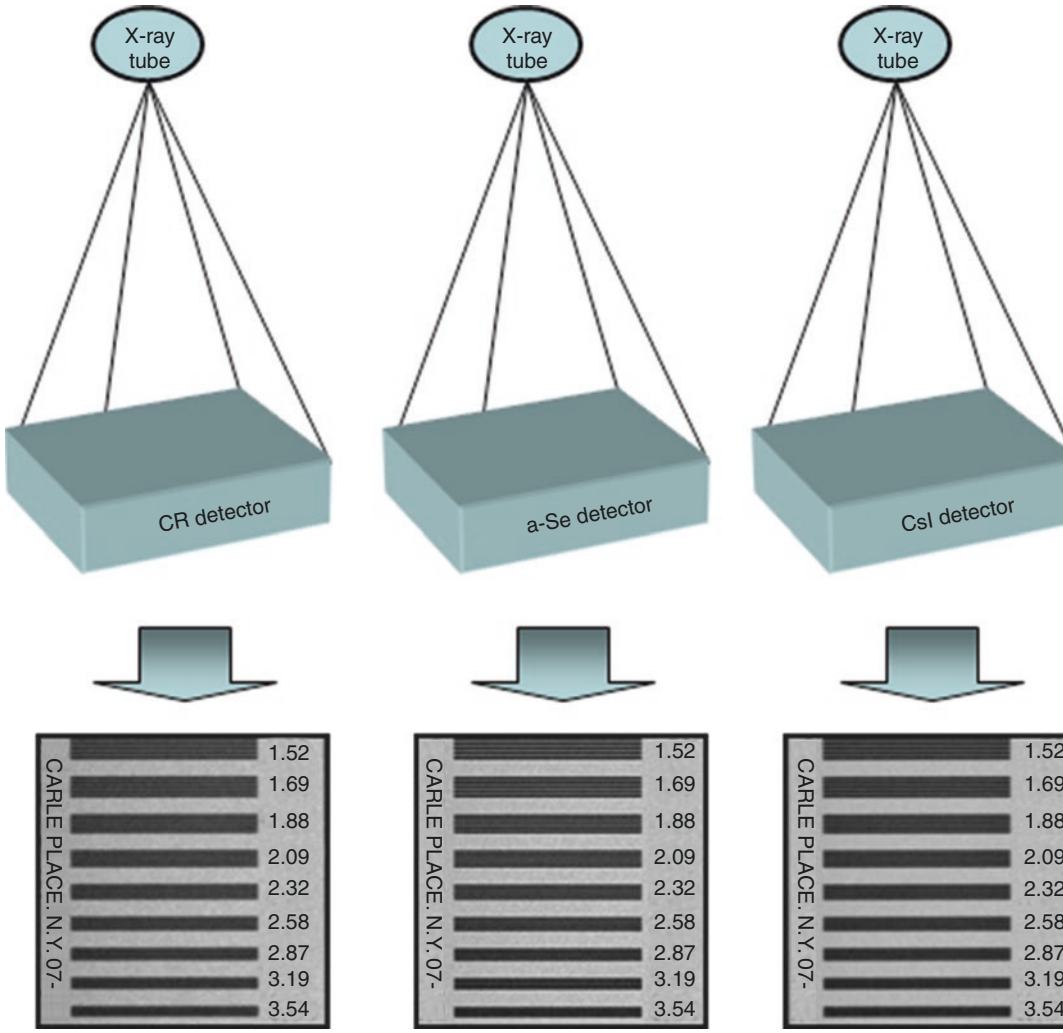

Fig. 4.19 The effect on image sharpness of a bar test pattern for three different digital radiography detector systems (Images of the test pattern provided by the kind courtesy of John Yorkston PhD, Senior Research Scientist, Clinical Applications Research, Carestream Health Inc. Rochester, NY)

Research at Carestream Health Inc., "these images demonstrate the "inherent" or "native" spatial resolution capabilities of the different technologies. Patient images typically get processed (potentially through proprietary algorithms) to optimize the visual presentation which can include varying degrees of "sharpening" (called a range of different names including "unsharp masking" for one). The success (or aggressiveness) of that step depends on the inherent signal to noise content of the image which is driven by the detector DQE" [10].

For the edged object and the narrow slit, an edge spread function (ESF) and a line spread function (LSF) have to be obtained, respectively. The LSF is shown in Fig. 4.20 for three digital radiography detectors. The narrower the LSF, the better the spatial resolution. It is clear that the spatial resolution is best with the a-Se detector and that structured CsI a-Si TFT detector produces better spatial resolution (narrower LSF) than the turbid Gd_2O_2S or CsI digital detector. The three methods listed above can be used to produce yet another function called the modulation transfer function (MTF).

4.8.2 Modulation Transfer Function

The *modulation transfer function (MTF)* is a complex mathematical function that measures the ability of the detector to transfer its spatial resolution characteristics to the image. Bushong [1] notes that the MTF can simply be expressed as a ratio of the image to the object and an MTF of 1 represents a perfect detector transfer. How is this MTF obtained?

Consider a patient's abdomen that consists of fine objects and coarse objects. These objects can be represented as spatial frequencies (line pairs/mm = lp/mm) where fine and coarse objects generate high and low spatial frequencies. While the

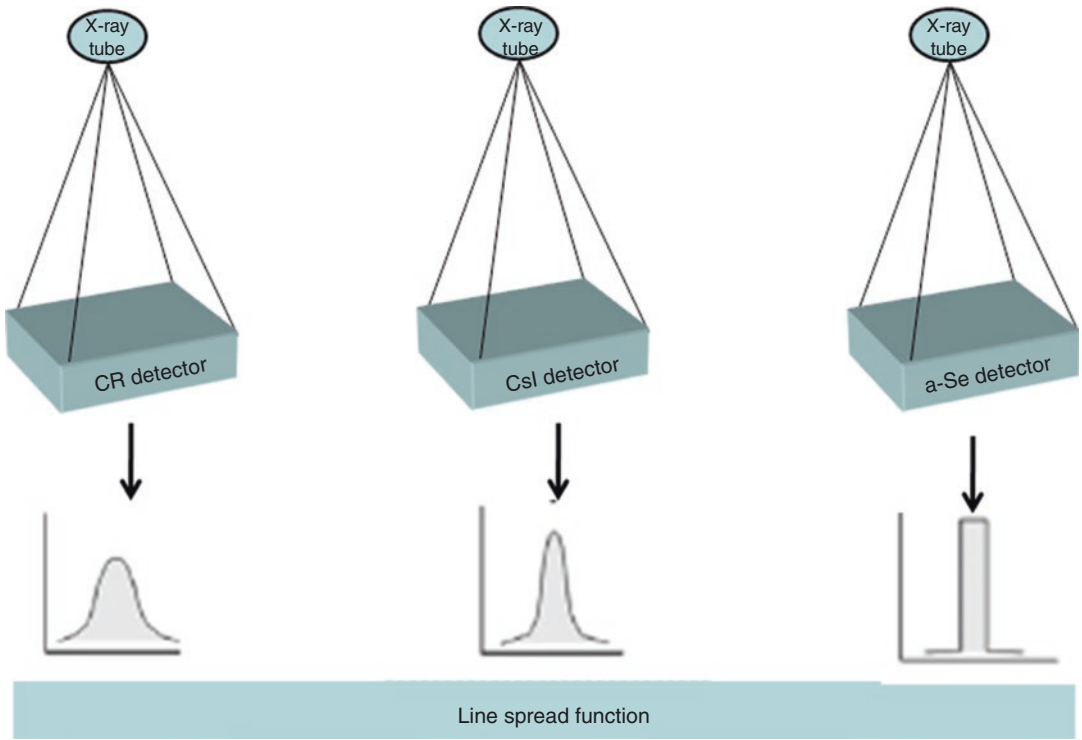

Fig. 4.20 The line spread function (LSF) for three different digital detector systems, CR, a-Si CsI structured phosphor TFT array, and a-Se TFT array (see text for further explanation)

high spatial frequencies represent fine detail or sharpness, the low spatial frequencies represent object contrast information. The image would contain both sharpness and contrast (intensity grayscale values). The MTF is a graph of the contrast plotted as a function of spatial frequency. As noted earlier an MTF of 1 represents a perfect transfer of spatial and contrast information.

Measurement of the MTFs for various detectors demonstrates that as the spatial frequency increases, the MTF decreases. A higher MTF value at a higher spatial frequency means that the detector provides better spatial resolution than lower MTF values at low frequencies. Furthermore, a higher MTF value at lower spatial frequencies means that the detector provides better contrast resolution.

All digital imaging systems have what is referred to as the limiting resolution, which is the spatial frequency limit that is obtained at an MTF value of 0.1. A system that has a higher spatial frequency at an MTF of 0.1 will show better spatial resolution than a system that has a lower spatial frequency at an MTF of 0.1.

4.8.3 Dynamic Range

The *dynamic range* of a digital detector is response of the detector to different levels of radiation exposure as is clearly illustrated in Fig. 4.10. As noted, the digital detector responds to a wider range of exposure compared with film-screen image receptors. This is one of the most obvious differences that allow the digital detector to respond to both underexposure and overexposure without the need for repeat exposures. It is important, however, to ensure that technologists use these detectors wisely in order to avoid "exposure creep." This was discussed earlier in the chapter, as well as in Chap. 3.

4.8.4 Detective Quantum Efficiency

The *detective quantum efficiency (DQE)* is yet another performance characteristic of digital detectors, and it was described in Chap. 3 for CR. Essentially, the DQE deals with how efficient a detector is at converting radiation falling upon it into a useful image signal. The DQE measurement is quite complicated and is obtained from other physical quantities such as the MTF, noise power spectrum (NPS), the incident X-ray flux, X-ray photon energy, and the spatial frequency (f).

The DQE provides information about the signal-to-noise ratio (SNR). In imaging, the goal is to obtain good contrast information in the image. This is, in general, the signal, and the detector must provide the maximum signal as possible. On the other hand, noise is also present in the signal due to few photons (quantum noise) striking the detector as well as system noise (electronic noise). These two parameters can be put together to create the SNR concept. In imaging applications, since contrast resolution (the ability to resolve small differences in tissue contrast) is mandatory for diagnostic interpretation, it is important to have a high SNR. This means that the signal is high and the noise is low.

The DQE can be expressed as follows:

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

A perfect digital detector would have a DQE of 1. Measurements of the DQE values for different digital radiography detectors used in clinical practice demonstrate that the indirect conversion CsI a-Si TFT flat-panel detectors offer the highest DQE at low frequencies. As the spatial frequencies increase, the DQE decreases rapidly [8].

A very important point about the DQE that the technologist should pay careful attention to is quoted from expert medical physicist Dr. Anthony Seibert, PhD [8], as follows:

> *A DR system with higher DQE, however, does not necessarily translate into a system with superior image quality or lower dose. For instance, a slot-scan digital system can achieve comparable image quality for chest imaging with a lower dose compared to a flat-panel large-area detector, despite having a lower intrinsic DQE(f). Besides intrinsic detector efficiency, other considerations affecting patient dose and image quality include effective*

X-ray energy, beam uniformity, wide-latitude response, spatial and temporal sampling, grid use, examination-specific image processing, image display quality, monitor calibration, viewing conditions, radiologist capability (alertness, experience), and working environment. The difference between a diagnostic and non-diagnostic image often has a cause other than the detector itself [8].

As mentioned earlier, the DQE also depends on the photon energy (kV). Under controlled conditions it has been shown that the quantum efficiencies (based on the X-ray absorption properties of the detector converter materials) at 70 kV is about 67% for a-Se and 77% for CsI phosphor. At 120 kV, efficiencies are 37% for a-Se and 52% for CsI. In lower kV applications such as mammography, a-Se performs better than CsI and has a higher DQE [11].

4.8.5 Image Lag

An important imaging performance characteristic of a flat-panel detector that has an impact on clinical practice is that of image lag [5, 11]. *Image lag*, also referred to as "memory effect" [11], refers to the persistence of the image, that is, charge is still being produced after the radiation beam from the X-ray tube has been turned off. The reason for this is that the "charge has been trapped in the metastable band-gap states in the a-Si and a-Se material during exposure and is only released slowly over time" [5]. Image lag times vary and are shortest for flat-panel digital detectors based on indirect conversion [11].

4.9 Image Artifacts

Flat-panel digital detectors are complex devices and pose numerous challenges in the manufacturing process. Flaws in the various components that make up the panel can lead to *image artifacts*. Several flaws, for example, include dust, scratches, chemical reactions of various materials that the detector is made of, and defective pixels [5]. Image artifacts can also arise from vibrations or poor performance of the electronics and scattered radiation grids as well. An example of a grid artifact (grid aliasing) is shown in Fig. 4.21.

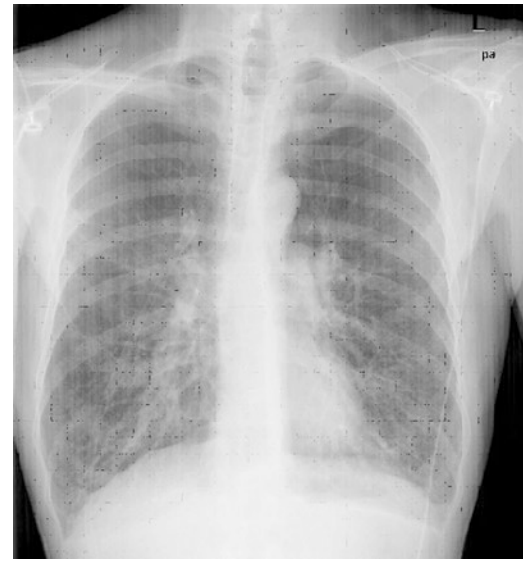

Fig. 4.21 Example of a grid artifact from a flat-panel digital detector (Courtesy of Kerry Krugh, PhD, Medical Physicist, University of Toledo Medical Center, Toledo, Ohio)

4.10 Other Applications of Flat-Panel Digital Detectors

Flat-panel digital detectors are used not only in general static radiography examinations in the main department but also in other imaging applications. Flat-panel detectors can be used in portable radiography applications and in applications involving fluoroscopy. Fluoroscopic applications include gastrointestinal tract fluoroscopy, angiography, portable fluoroscopy, and dedicated systems that have now become available for cardiology.

4.11 Wireless Flat-Panel Digital Detectors

Wireless flat-panel digital radiography (FPDR) became commercially available as of 2009 [12]. In this case the flat-panel detector is connected to the digital radiography console via a wireless

local area network (LAN) to send images and text communications (such as control commands from the console) to and from FPDR exposed detector to the digital radiography workstation. These activities are all executed in real time. Furthermore the FPDR detector has a built-in battery power supply in the form of a lithium ion capacitor as opposed to the lithium ion secondary battery, since this design reduces the any risk of injury to the patient [13]. The interested reader should refer to the various vendors for specific specifications of these devices. Specific specifications, for example, may include spatial resolution, sensitivity, wireless range, data transfer speed, details of battery use (e.g., charging time and charge life), and the control software features. In addition, readers who are interested in the DQE and the MTF for digital detectors should refer to a study by Samei et al. [14], for an initial exploration of these physical characteristics of wireless digital detectors.

In a study conducted by Garrido et al. [15], on the usefulness of wireless digital detectors in the emergency department of a hospital, the authors reported positive gains such as improved image quality and dose reduction to patients. Furthermore "the new system also has advantages in terms of functionality, ergonomics, and performance" [15].

As flat-panel detector techniques improve, they will become increasingly commonplace in the imaging environment and may replace other digital imaging systems such as CR systems, for example. Already, they are becoming more and more popular and are replacing the image intensifier tube used in fluoroscopy. Therefore the technologist must make every effort to have a good understanding of this technology. This chapter offers one small step in this direction.

References

1. Bushong S. Radiologic science for technologists. 11th ed. St Louis: Elsevier-Mosby; 2017.
2. Antonuk LE, Yorkston J, Huang W. A real time flat-panel amorphous silicon digital X-ray imager. Radiographics. 1995;15:993–1000.
3. Zhao W, Rowlands JA. X-ray imaging using amorphous selenium: Feasibility of a flat panel self-scanned detector for digital radiology. Med Phys. 1995;22(10):1595–604.
4. Korner M, et al. Advances in digital radiography: physical principles and system overview. Radiographics. 2007;27:675–86.
5. Yorkston J. Flat-panel DR detectors for radiography and fluoroscopy. In: Goldman L, Yester MV, editors. Specifications, performance, and quality assurance of radiographic and fluoroscopic systems in the digital era. College Park, MD: AAPM Monograph 30; 2004. p. 177–229.
6. Yorkston J. Digital radiographic technology. In: Samei E, Flynn M, editors. Advances in digital radiography. RSNA categorical course in diagnostic radiology physics. Oak Brook, IL: RSNA; 2003. p. 23–36.
7. Bushberg JT, Seibert JA, Leidholdt EM Jr, Boone JM. The essential physics of medical imaging. 3rd ed. Philadelphia: Lippincott Williams and Wilkins; 2012.
8. Seibert JA. Computed radiography/digital radiography: Adult. In: Frush DP, Huda W, editors. From invisible to visible-the science and practice of X-ray imaging and radiation dose optimization: RSNA categorical course in diagnostic radiology physics. Oak Brook, IL: RSNA; 2006. p. 57–71.
9. Willis CE. Computed radiography/digital radiography: pediatric. In: Frush DP, Huda W, editors. From invisible to visible-the science and practice of X-ray imaging and radiation dose optimization: RSNA categorical course in diagnostic radiology physics. Oak Brook, IL: RSNA; 2006. p. 78–83.
10. Yorkston J. Senior research scientist, clinical applications research, Carestream Health Inc. New York: Personal Communications; 2017.
11. Spahn M. Flat detectors and their clinical applications. Eur Radiol. 2005;15:1934–147.
12. Lanca L, Silva A. Digital imaging systems for plain radiography. New York: Springer Science+Business Media; 2013.
13. Tokuhiro O, Gidou T, and Kashino T. Development of a Wireless Cassette Digital Radiography Detector; Konica Minolta Medical & Graphic, Inc., Development Center, Development Department, Konica Minolta Technology Report Vol 8, 2011.
14. Samei E, Murphy S, Christianson O. DQE of wireless digital detectors: comparative performance with differing filtration schemes. Med Phys. 2013;40(8):081910.
15. Garrido Blazquez M, Agulla Otero M, Rodriguez Reco FJ, Torres Cabrera R, Hernando Gonzalez I. Wireless digital radiography detectors in the emergency area: an efficacious solution. Radiologia. 2013;55(3):239–46.

5 The Standardized Exposure Indicator

Abstract

This chapter addressed essential features of the international standardized EI for DR imaging systems. First, two "old" propriety scales for the EI, the inverse scale and the proportional scale, were reviewed briefly followed by an outline of a generalized method for determining the EI. The next topic described are three conditions for the IEC standardized EI that must be met: (a) the standardized EI is related to the detector exposure and the standardized EI is obtained from the pixel values in the region of interest, (b) the standardized EI uses a linear proportional scale related to the detector exposure/signal (i.e., doubling the detector dose doubles the standardized EI value), and (c) consideration of the radiation beam quality (kVp, half-value layer, and added filtration) used for the calibration of the EI and the precision of the scale. Additionally, selected terminology of the IEC standardized EI are defined and include the exposure index (EI), the target EI (EI_T), deviation index (DI), and the value of interest (VOI).

The IEC standardized EI is now proportional to the detector exposure and requires the user to establish EI_T values for all examinations in order to ensure optimization of the dose to the patient without compromising the image quality. These values (EI and EI_T) can now be used to calculate the DI. The DI provides immediate feedback to the technologist as to whether the correct exposure was used for the examination. The chapter concludes with an example of a dose-image quality optimization study for the purpose of providing insights as to how a department may go about establishing EI_T values objectively.

It is important to note that sections of this chapter have been previously published in *Journal of Medical Imaging and Radiation Sciences* 45 (2014) 144–158, entitled "The New Exposure Indicator for Digital Radiography," and some of the content has been taken from my PhD thesis, entitled *Optimization of the Exposure Indicator as a Dose Management Strategy in Computed Radiography*. PhD Dissertation (Charles Sturt University, New South Wales, Australia, 2014).

5.1 Introduction

In Chaps. 3 and 4, the exposure response of film-screen (FS) radiography was outlined. In summary, the response describes a curve referred to as the characteristic curve which shows how the optical density of the film responds to the log of the relative exposure. While high exposures produce dark images (overexposed), low exposures result in light images (underexposed) that are not diagnostic and therefore have to be discarded. Acceptable images are produced with exposures that fall within the slope of the curve. The light

E. Seeram, *Digital Radiography*, https://doi.org/10.1007/978-981-13-3244-9_5

and dark images are used as an exposure indicator by the technologist to indicate whether the correct exposure technique factors have been used for the examination under study. A major problem with FS radiography is the narrow exposure latitude, which means that careful attention must be paid in the selection of the correct exposure technique factors (mAs and kV). This problem is overcome by digital radiography (DR). Since DR systems have a linear response to the exposure, the advantage is that of wide exposure latitude that is 100 times that of FS radiography. Additionally, the image processing afforded by digital radiography renders consistent appearance of images that are overexposed and underexposed, as illustrated in Fig. 5.1.

The graphs illustrate histograms and frequency distributions of the digital code values and the characteristic curves that are processed to convert the digital values into images of consistent image quality. As noted by Seibert and Morin [1], "the VOILUT is adjusted to the histogram to achieve optimal rendering of the image contrast." The problem of this operation is that is that the user now has difficulty in identifying images that have been underexposed and overexposed. In attempting to produce acceptable images users appear to increase exposure technique factors which ultimately result in what has been referred to as "exposure creep" or simply "dose creep" [1–3].

In DR systems, vendors address these problems by providing what they referred to as an exposure indicator (EI), sometimes called an exposure index, to indicate whether the correct exposure technique factors have been used for the examination. Various DR vendors in the past have used different EIs with specific names [4], as described in detail in Chaps. 3 and 4. The fundamental problem with different nomenclature for the EI is the fact that users are often confused and frustrated during routine use of different DR systems in the department [1]. To address this problem, the International Electrotechnical Commission (IEC) [5] and the American Association of Physicists in Medicine AAPM [6] have developed a standardized EI for use with DR systems.

This chapter outlines the essential features of the standardized exposure indicator through a brief review of the "old" EI. In addition, the chapter concludes with an insight into optimization research using the standardized EI characteristics.

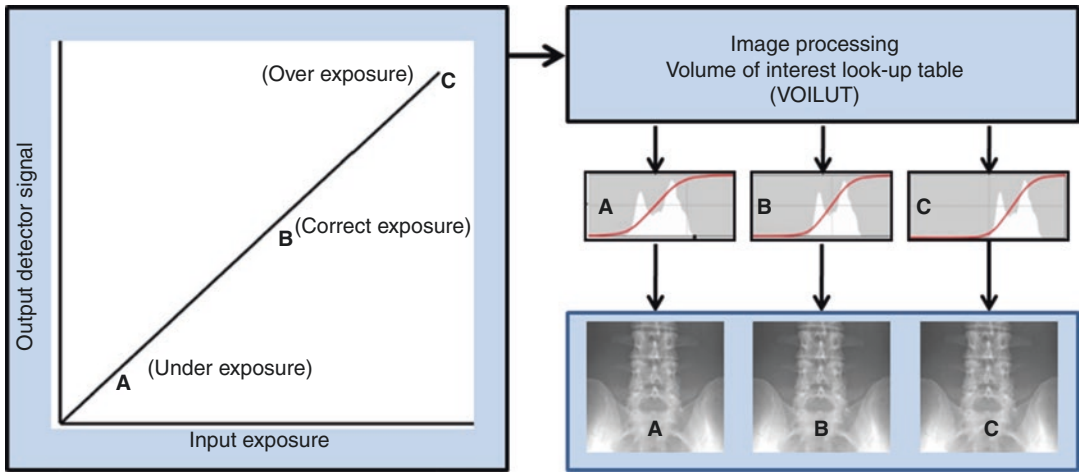

Fig. 5.1 In digital radiography, image processing allows for consistent appearance of images that are overexposed and underexposed. See text for further explanation

5.2 Propriety EI Scales: A Brief Review

In Chaps. 3 and 4, the exposure indicator (EI) was described. In review, the EI does not refer to patient dose, but rather the EI is linked to the exposure to the digital detector. The IEC defines the EI for digital radiography as "a measure of the detector response to radiation in the relevant region of an image" [5]. Figure 5.2 illustrates two different approaches (or scales) to express the EI as a function of the radiation dose. While the first is an inverse scale which explains that as the dose increases the EI decreases, the second is a proportional scale which shows that as the dose increases, the EI increases proportionally.

5.3 Determination of the EI

The procedure for calculation of the EI is shown in Fig. 5.3. First the digital detector is exposed, and the dose distribution (usually in mGy) is used by the computer to produce a digital image from this dose. Subsequently the EI is obtained from the information in the image by using one of two methods, the histogram method, or the region of interest (ROI) method. A procedure referred to as the VOI (pixel value of interest) segments the anatomy under study, and an average value is obtained from at least 25% of the central region of the image [3]. If the histogram method is applied, the mean value of the relevant range of pixel values is applied in calculating the EI. Since several DR systems are available from different vendors, they each have their proprietary approaches for deriving the EI value.

5.4 The Standardized EI

The problems of different proprietary methods and different approaches to obtaining the EI have resulted in a need for clarity with regard to the calculation of the EI [7]. This task was first identified in 2004 at the as low as reasonably achiev-

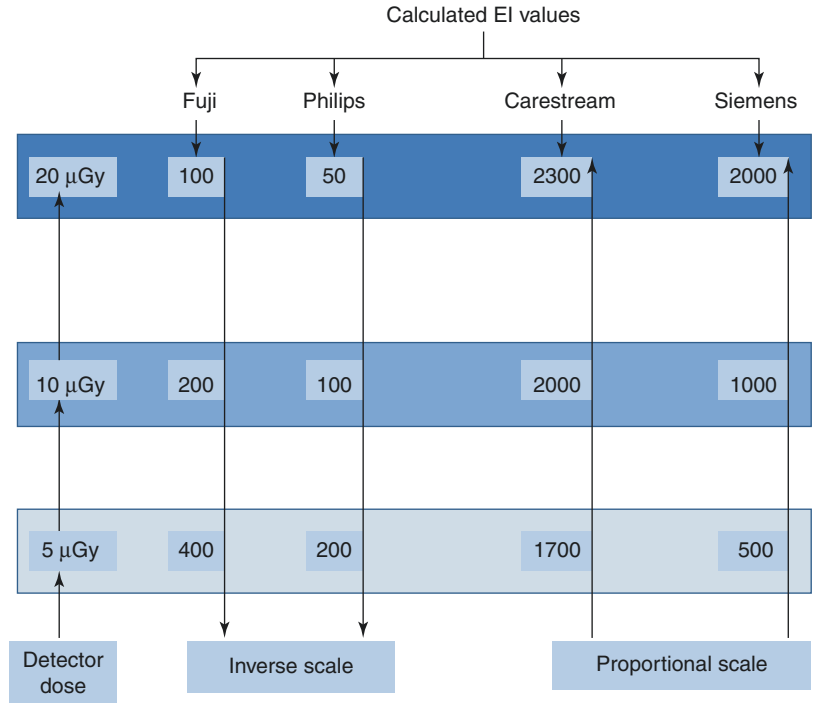

Fig. 5.2 Two different approaches (or scales) to express the EI as a function of the radiation dose. While the first is an inverse scale which explains that as the dose increases the EI decreases, the second is a proportional scale which shows that as the dose increases, the EI increases proportionally

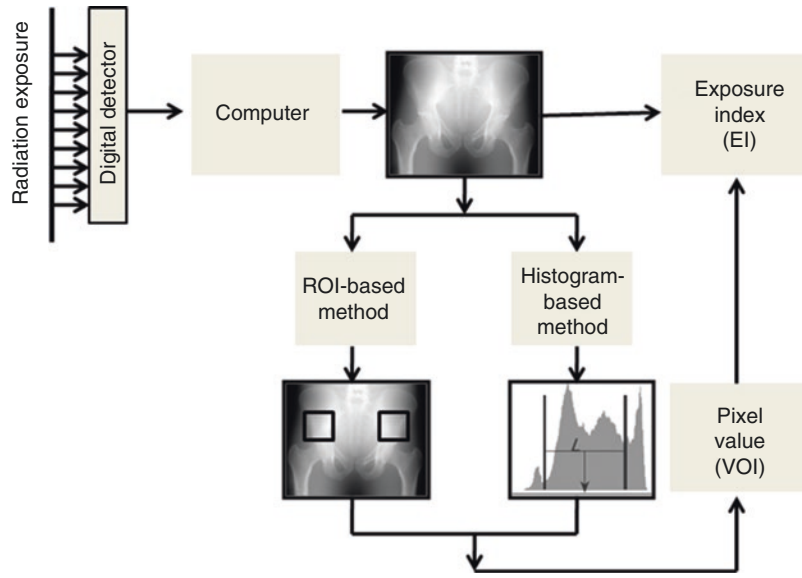

Fig. 5.3 Two popular methods to calculate the EI include the ROI-based method and the histogram-based method. See text for further explanation

able (ALARA) special dose conference on the use of CR and flat-panel DR systems in pediatric radiology [8]. In 2006, Van Metter and Yorkson [9] recommended that a more generalized and simple metric that does not depend on the digital radiography technology and the vendor.

In an effort to address this need, the IEC, the AAPM, and the Medical Imaging and Technology Alliance (MITA) worked on establishing a standardized EI for DR. The IEC described their approach to developing a standardized EI in 2008 [5]; the AAPM described their methodology in the *Journal of Medical Physics* in 2009 [6]. The titles for these two significant reports are shown in Fig. 5.4. As a result, DR vendors have implemented these standards.

5.4.1 Conditions for the IEC Standardized EI

In establishing a standardized EI for DR, the IEC requires that the following elements be met: First, a direct proportional approach or scale must be used which relates the EI to the detector dose (i.e., doubling the detector dose doubles the standardized EI value). Second, the ROI pixel values are used in calculation of the standardized EI. Furthermore the IEC specifies that other factors must be considered as well in establishing a standardized EI such as the kV, filtration, and so forth.

5.4.2 Definitions

The IEC [5] has defined four terms that relate to the standardized EI that are important when using the EI in clinical practice. These are exposure index (EI), the target exposure index (EI_T), the deviation index (DI), and the value of interest (VOI). While the EI is a "measure of the detector response to radiation in the relevant image region of an image acquired with a digital X-ray imaging system" [5], the EI_T is the "expected value of the exposure index when exposing the X-ray image receptor properly" [5]. The DI on the other hand is a "number quantifying the deviation of the actual exposure index from a target exposure index" [5]. Finally the IEC states that the VOI is the "central tendency of the original data in the relevant image region. The central tendency is a statistical term depicting generally the center of a distribution. It may refer to a variety of measures such as the mean, median, or the mode." The VOI is important for the calculation of the EI.

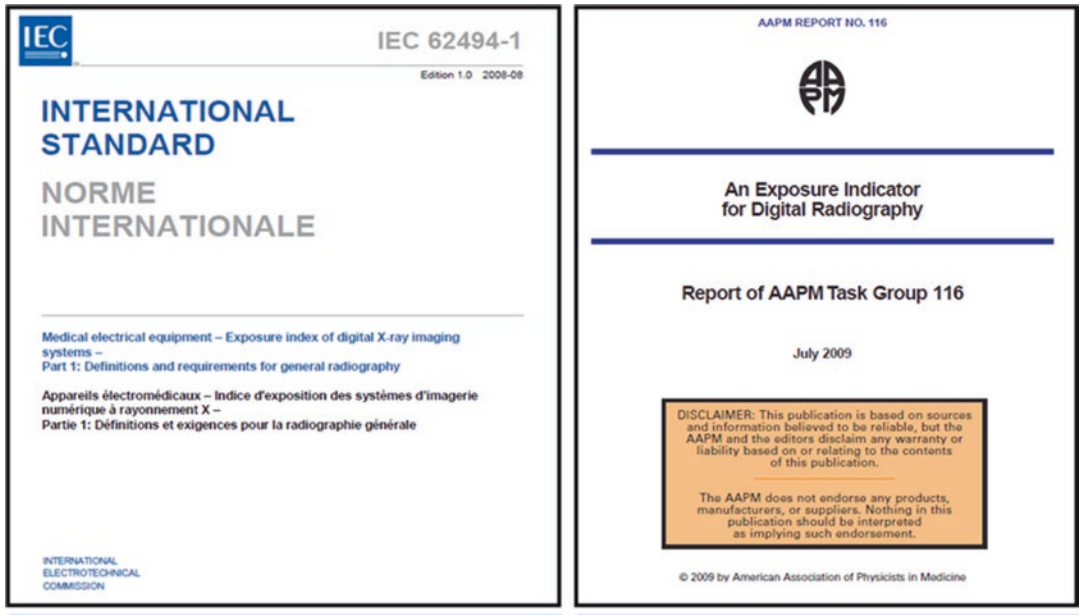

Fig. 5.4 Titles for two significant reports describing the approaches to standardize the EI in digital radiography, that of the IEC and the AAPM

5.4.3 Essential Steps to Determine the Standardized EI

Both the IEC and the AAPM suggest the following steps to obtain the standardized EI. First, the detector must be calibrated such that there is a direct proportional relationship between the EI and the exposure at the detector, that is, if the detector exposure is doubled, the EI value doubles. For example, detector exposures of 2 μGy, 4 μGy, 8 μGy, and 10 μGy produce EI values of 200, 400, 800, and 1300, respectively [3]. Figure 5.5 illustrates an example of the IEC standardized EI values for the Philips Healthcare DR system. In addition, users must note that the standardized EI depends on X-ray beam quality (kVp). For example, for CR, a dose of 10 μGy at 70 kVp and 90 kVp will generate EI values of 1000 and 800, respectively [11, 12].

The calibration procedure will not be described here since it is comprehensive; however, such procedure is described by Shepard et al. [10]. Secondly, the EI_T should be determined by the department (not the vendor) using objective approaches. For the third essential task, the DI values can be computed using the following mathematical relationship:

$$DI = 10 \log_{10} \left(EI / EI_T \right)$$

5.4.4 Interpretation of DI Values

The value of the DI obtained indicates how far the actual EI value is different from the user established EI_T values [1, 11, 13]. Figure 5.6 provides ranges of DI values that are significant in helping the technologist determine acceptable and unacceptable images [1, 11]. The DI is an essential tool that indicates that the correct exposure technique factors were for the body part under investigation. Furthermore, Fig. 5.6 provides a snapshot of the meaning of various DI values to guide the user in how to use this information effectively in clinical practice, thus

Fig. 5.5 An example of the IEC standardized EI values for the Philips Healthcare DR system

Radiation dose to the detector (μGy)	Old proprietary EI values (Philips healthcare)	Standardized IEC EI values (Philips healthcare)
1.25	500	125
2.5	400	250
5	200	500
10	100	1000

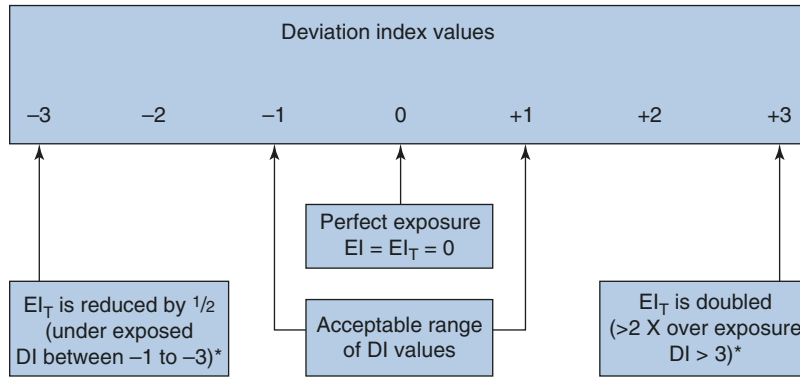

Fig. 5.6 Ranges of DI values that are significant in helping the technologist to determine perfect, acceptable, and unacceptable images

removing the phenomenon of "exposure creep" [14]. In addition, the AAPM [6] provides "actions" that should be taken for each of the DI values shown in Fig. 5.6. For DI greater than +3 (excessive radiation exposure) and from +1 to +3 (overexposure), the action is to "repeat only if relevant anatomy is clipped or 'burned out.'" When the DI value is less than −1 (underexposed), the action is to "consult radiologist for repeat" and when the DI value is less than −3, the action is to repeat.

5.4.5 Factors Affecting EI Values

There are several other factors affecting the EI other than the exposure, and these have been placed into two categories, namely, human factors and technique factors [3]. While the former includes collimation, image cropping, and positioning of the patient, the latter includes determination of the "relevant image region," the use of automatic exposure control (AEC) as opposed to manual exposure timing, and the X-ray beam quality [3].

5.5 The Standardized EI: Responsibilities

The effective use of the international standardized EI places certain responsibilities on both the vendor and the user. While the vendor has the responsibility of ensuring technical details of EI

calibration and specific image processing operations, using defined methodology, they also establish EI_T values before the equipment is shipped to the user [11]. A significant responsibility is for users to establish their own EI_T values based on the exposure technique factors used in their departments for different body parts, etc. This task must be conducted by using objective measures. There are other responsibilities that should be considered as well. For example, when the department is considering new equipment, it is important that they "insist on the implementation of the EI standard, IEC 62494-1." Furthermore, Seibert and Morin also suggest that an important responsibility should also focus on "training technologists on the use of the DI, acceptable limits of the DI value, and compensation methods when the DI is beyond the acceptable range" [1].

References

1. Seibert JA, Morin RL. The standardized exposure index for digital radiography: an opportunity for optimization of radiation dose to the pediatric population. Pediatr Radiol. 2011;41:573–81.
2. Gibson DJ, Davidson R. Exposure creep in CR: a longitudinal study. Acad Radiol. 2012;19:458–62.
3. Neitzel U. Ending a state of confusion: NACP optimization course. Hamberg: Philips healthcare, diagnostic X-ray; 2013.
4. Brake DA. A standardized exposure index for digital radiography. Radiol Technol. 2016;87(5):581–5.
5. International Electrotechnical Commission. IEC 62494-1 ed. 1 Medical electrical equipment exposure index of digital X-ray imaging systems, part 1: definitions and requirements for general radiography, Geneva: IEC; 2008. p. 2008.
6. American Association of Physicists in Medicine (AAPM) Report No 116. An exposure indicator for digital radiography. College Park, MD: American Association of Physicists in Medicine; 2009.
7. Bushberg J, Seibert AJ, Leidholdt M, Boone JM. The essential physics of medical imaging. 3rd ed. Philadelphia: Lippincott Williams & Wilkins; 2012.
8. Willis CE, Slovis TL. Conference on the ALARA (as low as reasonably achievable) concept in pediatric CR intelligent dose reduction. Pediatr Radiol. 2004;34(Suppl. 3):S159–247.
9. Van Metter R, Yorkston J. Toward a universal definition of speed for digitally acquired projection images. In: Flynn MJ, Hsieh J, editors. Medical imaging 2006: physics of medical imaging. San Diego, CA: Society of Photo-optical Instrumentation Engineers; 2006. p. 442–25.
10. Shepard JS, Wang J, Flynn M. An exposure indicator for digital radiography: AAPM task group 116 (executive summary). Med Phys. 2009;36(7):2898–914.
11. Don S, MacDougall R, Strauss K, et al. Image gently back to basics initiative: ten steps to help manage radiation dose in pediatric digital radiography. Am. J Roentgenology. 2013;200(5):W431–6.
12. Seeram E, Davidson R, Bushong S, Swan H. Optimizing the exposure indicator as a dose management strategy in computed radiography. Radiol Technol. 2016;87(4):380–91.
13. Moore Q, Don S, Goske M. Image gently: using exposure indicators to improve pediatric digital radiography. Radiol Technol. 2012;84(1):93–9.
14. Seeram E. The new exposure indicator for digital radiography. J Med Imaging Radiat Sci. 2014;45:144–15.

Digital Fluoroscopy

6

Abstract

Fluoroscopy is an imaging modality that shows anatomical structures and the motion of organs and the movement of contrast media in blood vessels and organs with the goal of obtaining functional information. It has evolved from conventional fluoroscopy recording images on film to current digital fluoroscopy (DF) in which digital dynamic images are obtained and stored in a computer. There are two modes of DF, one that is based on using an image intensifier coupled to a digital imaging chain and one that is based on the use of digital flat-panel detectors (FPDs). The former is being replaced by the latter, which is capable of producing dynamic images that can be displayed and viewed in real time. For this reason, these detectors are sometimes referred to as dynamic FPDs. Two types of dynamic FPDs are currently available for digital fluoroscopy, namely, the cesium iodide amorphous-silicon (CsI a-Si) TFT indirect digital detector and the a-selenium TFT direct digital detector. An important characteristic of these detectors is that they must have high frame rates and fast data transfer rates. Frame rates of 15–30 frames per second (fps) or greater are possible at readout speeds of 30–50 ms. Additionally, dynamic FPDs can operate in at least two readout modes: the fps in continuous X-ray mode and the fps in the pulsed X-ray mode. The overall goal of the latter is to reduce the dose to the patient during the examination.

FPD DF systems offer two approaches to image magnification: (a) electronic magnification (zoom) and (b) binning. With the former approach, there is no increase in spatial resolution, and both original and magnified images have the same signal-to-noise ratio (SNR). Binning has the disadvantage of less spatial resolution because the effective area of each image pixel is four times larger, and it has the advantage of lower data rates and less image mottle than ungrouped pixels. Advantages of FPD DF include distortion-free images, improved contrast resolution, high detective quantum efficiency (DQE), and uniform image quality over the whole displayed rectangular image. The chapter concludes with a brief description of several image post-processing operations specifically for digital fluoroscopy, such as grayscale image manipulation, temporal frame averaging, last image hold, and edge enhancement and a brief overview of digital subtraction angiography (DSA).

6.1 Introduction

In Chaps. 3 and 4, digital radiographic imaging systems, namely, computed radiography (CR) and flat-panel digital radiography (FPDR), were described in detail. While these systems produce static or stationary images, fluoroscopy is an

E. Seeram, *Digital Radiography*, https://doi.org/10.1007/978-981-13-3244-9_6

imaging modality that produces dynamic or moving images, displayed in real time. The purpose of fluoroscopy is to study not only anatomical structures but also, more importantly, the motion of organs and the movement of contrast media in blood vessels and organs with the goal of obtaining functional information.

The introduction of fluoroscopy dates back to 1896, soon after the discovery of X-rays in 1895, when Thomas Edison developed the first fluoroscope [1] that allowed for the observation of moving images such as the beating heart, in real time, for example. Through the years, fluoroscopy has developed into a sophisticated imaging system as new technologies emerged to improve its imaging system characteristics. For example, the image intensifier tube was developed to solve the image quality problems imposed by the fluoroscopic screen of the early fluoroscopes. A brief timeline of the introduction of several technologies for fluoroscopic imaging is shown in Fig. 6.1. It is clear that fluoroscopy has evolved from analog imaging systems to digital imaging systems that use flat-panel digital detectors.

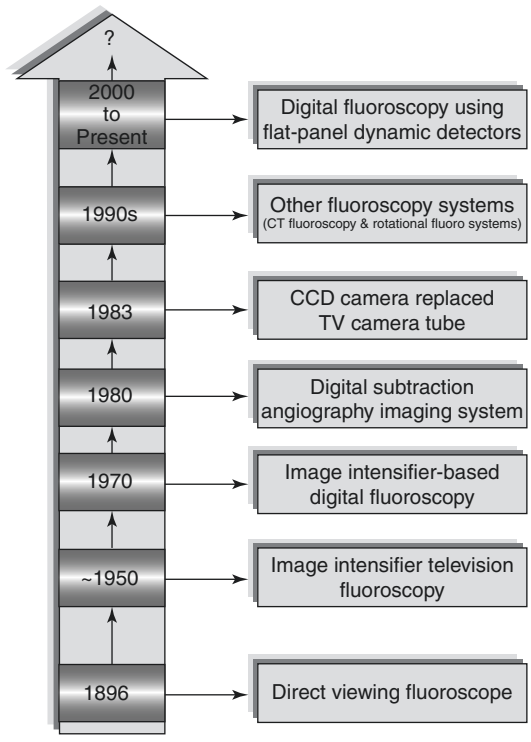

Fig. 6.1 A brief timeline of the introduction of different technologies for fluoroscopic imaging

The purpose of this chapter is to review the physical principles and technology of conventional fluoroscopy and describe the unique features that characterize digital fluoroscopy. These features include the conversion of analog data from the image receptor (image intensifier and/or flat-panel digital detectors) into digital data. The digital data is subsequently processed by a computer to produce digital fluoroscopic images.

6.2 Conventional Fluoroscopy Principles: A Review

The term *conventional fluoroscopy* refers to the use of an image intensifier coupled to a video camera that converts the image from the output screen of the image intensifier into a video signal (analog data). This signal is sent to a television monitor where images are displayed at frame rates of at least 30 frames per second (fps) to provide the effect of motion [1–5]. The next section of the chapter will review the physical principles and major technical components of a conventional fluoroscopic imaging system and its associated image quality characteristics.

6.2.1 Imaging Principles and Technical Components

A typical conventional fluoroscopic imaging chain is shown in Fig. 6.2. The major technical components include the X-ray tube and generator, the spot-film device, the image intensifier tube, the optical image distributor, the photospot camera, and a video camera coupled to an X-ray television display monitor.

While the *X-ray tube and generator* provide the appropriate X-ray beam for both fluoroscopy and radiography, the *image intensifier converts* X-rays into light that is captured by the video camera. The output video signal from the video camera goes to the television display monitor to create fluoroscopic images. Single static images can be recorded by using the spot-film device

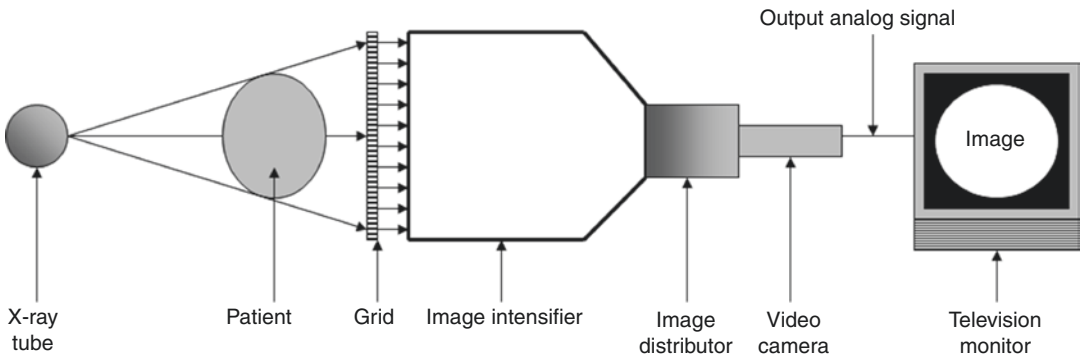

Fig. 6.2 The technical components of a typical conventional fluoroscopic imaging chain

(in which case the system must switch from fluoroscopic exposure technique factors to radiographic factors) and/or the photospot film camera. The transition to digital fluoroscopy has rendered these two methods that are almost obsolete since they both use film for recording images. Another important element of a conventional fluoroscopic imaging system is closed-circuit X-ray television. This system couples the video camera to the television monitor by means of a coaxial cable and control electronics. It is important to note that the video camera can be either a television "pickup" tube or a charge-coupled device (CCD). In modern fluoroscopic systems using image intensifiers, the CCD has replaced the television camera tube.

6.2.1.1 X-Ray Tube and Generator

The X-ray tube used in fluoroscopy must be capable of producing X-rays either continuously or in short bursts or pulses (pulsed fluoroscopy). While the former allows for real-time image display (30 fps at 33 ms), the overall goal of the latter is to reduce patient dose especially in pediatric fluoroscopy. In pulsed systems, pulses are as short as 3–10 ms/image [1, 5]. Furthermore, pulsed fluoroscopy requires the use of a grid-controlled X-ray tube, and the dose can be reduced by as much as 90% compared to non-pulsed fluoroscopy [1]. Additionally, since the pulses are very short (3–10 ms), there is less blurring of moving structures.

Modern conventional fluoroscopy imaging units use a high-frequency generator to ensure efficient production of X-rays. In addition, fluoroscopic exposure techniques utilize low mA and high kV exposure factors. For example, mA values from 1 to 3 mA and kV values from 65 to 120 kV are not uncommon. For recording images using the cassette-loaded spot-film device, radiographic factors are used. This means that the imaging system generator must be capable of switching from fluoroscopic mode to radiographic mode. In the latter mode, the tube current is increased to higher mA values [1].

6.2.1.2 The Image Intensifier Tube

The image intensifier tube has been developed to replace the conventional fluorescent screen of the early fluoroscopes. An early fluoroscope is graphically illustrated in Fig. 6.3. The screen was made of zinc cadmium sulfide (ZnCdS) and emitted a yellow-green light when struck by X-rays. A major problem with this screen was that the image lacked detail, contrast, and brightness. The ability to perceive detail is limited by geometric factors, rod vision, and the low brightness levels of the screen. Efforts to solve these problems only resulted in greater dose to patients. Additionally, early fluoroscopy had to be performed in the dark using red goggles for the purpose of dark adaptation. The above problems have been solved with the image intensifier through a process referred to as image intensification.

Image intensification refers to the brightening of the fluoroscopic image using the image intensifier. The structure of the image intensifier tube

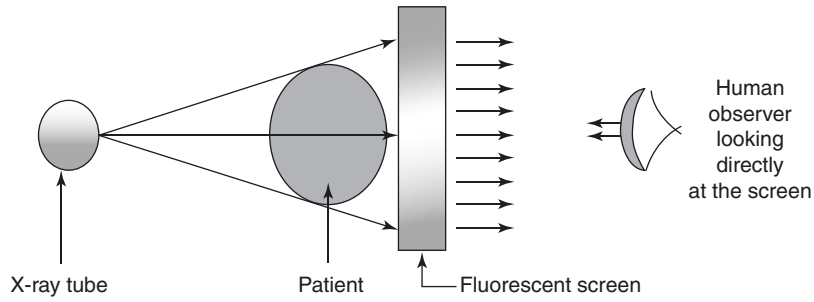

Fig. 6.3 The fluoroscope was the first system for the direct observation of moving structures in the patient's body. The images are produced on a fluorescent screen that emits light when struck by X-rays

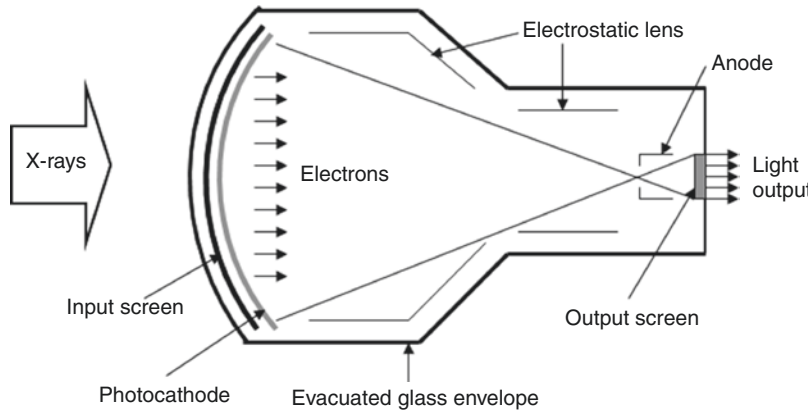

Fig. 6.4 The major components of an image intensifier tube. See text for further explanation

is shown in Fig. 6.4. These include the input screen, the photocathode, the electrostatic lens, and an output screen, all enclosed in an evacuated glass envelope.

The input screen is coated with a phosphor that converts X-ray photons to light photons. The state-of-the-art phosphor is cesium iodide (CsI). The CsI phosphor absorbs twice as much radiation compared with ZnCdS and is packed in a needlelike fashion (structured phosphor as described and illustrated in Chap. 4) to reduce the lateral spread of light in an effort to improve the spatial resolution compared with powdered CsI phosphors. The diameter of the input screen is variable; however, diameters ranging from 13 cm to 30 cm are not uncommon. Larger-diameter image intensifiers (36 –57 cm) have become available for imaging larger anatomical regions such as the abdomen.

The light from the input screen (phosphor) strikes the photocathode which is made of antimony cesium (SbCs) and which emits photoelectrons. At an energy of 60 keV, one X-ray photon at the input screen will result in about 200 photoelectrons being emitted at the photocathode [6]. Multialkali photocathodes with a combination of potassium, sodium, and cesium will emit about three times more photoelectrons than SbCs photocathodes, making these image intensifiers much more efficient than single alkali photocathodes.

The electrostatic lens or electron optics as it is often referred to consists of a series of electrodes that accelerate and focus the photoelectrons from the photocathode to the output screen. This requires a voltage of about 25–30 kV applied between the photocathode and the output screen.

The output screen is coated with ZnCdS phosphor that converts the photoelectrons into light. The diameter of the output screen is about one-tenth the diameter of the input screen. Due to the acceleration of the photoelectrons and the small size of the output screen, the image at the output screen is extremely bright. This increase in brightness is conveniently referred to as

brightness gain (BG). BG can be obtained using the following relationship:

$$\text{BG} = \text{minification gain}\left(\text{MG}\right) \times \text{flux gain}\left(\text{FG}\right)$$

The MG is a ratio expressed as follows:

$$\text{MG} = \left[\frac{\text{Diameter of the input screen}}{\text{Diameter of the output screen}}\right]^2$$

The FG on the other hand is also a ratio expressed as follows:

$$\text{FG} = \frac{\text{Number of light photons at the output screen}}{\text{Number of light photons at the input screen}}$$

The brightness gain concept has been replaced by another method used to measure the intensification of the image intensifier tube. This is the conversion factor (CF) which measures the light gain at the output screen using the following relationship:

$$\text{CF} = \frac{\text{Luminance of the output screen}}{\text{Exposure rate at the input screen}}$$

While the unit of luminance (light brightness) is the candela/square meter (Cd/m^2), it is milliroentgens/second (mR/s) for the exposure rate. Bushong [1] reports that while the brightness gain can range from 5000 to 30,000, the conversion factor for image intensifiers can range from 50 to 300. A higher conversion factor implies that the intensifier is much more efficient than the one with a lower conversion factor.

6.2.1.3 Image Intensifier Tube Housing

The image intensifier glass envelope is enclosed in a metal housing which not only provides mechanical support for the glass envelope but also shields the intensifier against magnetic fields. Since the housing is lined with lead, it also shields from any radiation scattered within the glass envelope.

6.2.1.4 Optical Image Distributor

The position of the optical distributor is shown in Fig. 6.2. The purpose of the image distributor is to split the total light (100%) from the output screen between the video camera and the photospot film camera. Using a system of lenses and a beam-splitting mirror, 10% of the light goes to the video camera, and 90% of the light goes to the photospot film camera.

6.2.2 Magnification Fluoroscopy

Magnification of the image in conventional fluoroscopy is an important feature of the image intensifier. The purpose of magnification fluoroscopy is to enhance the image in order to facilitate diagnostic interpretation.

Magnification fluoroscopy is only possible with multi-field image intensifiers. These include the popular dual-field and the triple-field intensifiers that use a technique referred to electron optical magnification [1]. This technique changes the voltage on specific electrodes of the electrostatic lens system in the image intensifier tube, to cause the electron beam crossover point to increase its distance from the output screen. A dual-field intensifier (25 cm/17 cm) can operate in the full-field mode (25 cm) and in the magnification mode (17 cm). When the magnification mode is used, the X-ray beam is automatically collimated to fall upon the central portion of the input screen to cover a diameter of 17 cm. On the other hand, a triple-field intensifier (25 cm/17 cm/12 cm) can operate in two magnification modes, 17 cm and 12 cm modes. The 12 cm mode provides greater magnification than the 17 cm mode.

Magnification provides increased spatial resolution but at the expense of increased dose to the patient. In general the increase is about 2.2 times that used in the full-field mode of operation. For example, the exposure rate for a 25 cm mode is about 30 μR/s, while it is 60 μR/s and 120 μR/s for the 17 cm and 12 cm modes, respectively [6]. The approximate dose can be computed by using the ratio of the two fields as follows:

$$\text{Dose} = \frac{\left(\text{Full-field diameter}\right)^2}{\left(\text{Magnification mode}\right)^2}$$

For example, the dose increase when going from a 25 cm mode of operation to a

magnification mode of 12 cm is about 4.3 times ($25^2/12^2 = 4.3$).

6.2.3 Image Quality Characteristics

There are at least three image quality parameters of the image intensifier that are worthy of review in this chapter. These include spatial resolution, contrast ratio, and noise. In addition, image intensifiers do exhibit few artifacts, such as image lag, vignetting, pincushion distortion, and "S" distortion, as well as veiling glare [1, 2]. Each of these will now be highlighted briefly.

6.2.3.1 Spatial Resolution

The *spatial resolution* of an image intensifier refers to its ability to resolve fine details in an object (patient). Since the input screen is convex with respect to the X-ray tube (see Fig. 6.4), the spatial resolution is much better at the center of the input screen compared to the screen's periphery. While the spatial resolution for CsI image intensifier operating in the 25 cm mode is 4 line pairs/mm (lp/mm), it is 6 lp/mm in the 10 cm mode. This means that 0.125 mm objects can be visualized for the 4 lp/mm- and 0.08-mm-sized objects can be seen when the resolution is 6 lp/mm [1].

6.2.3.2 Contrast Ratio

The *contrast ratio* of an image intensifier tube is the ratio of the image brightness at the periphery to that of the center of the output screen. A typical contrast ratio is about 20:1, but this can range from 10:1 to 30:1 and 15:1 to 35:1 depending on where it is measured [6].

6.2.3.3 Noise

Conventional fluoroscopy with an image intensifier operates in the low mA mode, and therefore the noise level is usually high. To reduce the *noise*, the mA can be increased but results in a proportional increase in patient dose. To reduce this dose, the ZnCdS input phosphor used in the early image intensifier tubes was replaced with CsI input phosphor, since CsI has a higher quantum detection efficiency (QDE) and uses very little radiation dose to produce good quality images that are noise-free.

6.2.3.4 Image Intensifier Artifacts

Image intensification fluoroscopy can produce several *artifacts* mentioned earlier. Image lag is the continued emission of light from the screen when the radiation beam has been turned off. This is not a serious problem however with newer image intensifiers, since the lag time is in the order of about 1 ms [2, 6]. Vignetting refers to a loss of brightness at the periphery of the image. This means that the image is sharper and much brighter in the central portion of the screen. The image intensifier may also exhibit veiling glare, an artifact that results when light is scattered in the intensifier tube.

Two other image intensifier artifacts include pincushion distortion and "S" distortion as illustrated in Fig. 6.5. When a rectangular grid is image with an intensifier, then pincushion distortion (Fig. 6.5a) results due to the fact that the input screen is curved. On the other hand, "S" distortion (Fig. 6.5b) appears if an electromagnetic field is close to the intensifier. This field will influence the electrons especially those at the "the perimeter of the image intensifier more so than those nearer the center" [6].

6.2.4 Fluoroscopic Television Chain

The image at the output screen of the image intensifier tube is far too small and too bright to be observed directly by a radiologist. Therefore a *closed-circuit television chain* is used to display this image onto a television monitor for proper viewing and interpretation. The technical components of a fluoroscopic television chain are shown in Fig. 6.6. These include a video camera and a television monitor coupled by a cable referred to as a coaxial cable.

The video camera can be a television "pickup" camera tube or a CCD camera, and it is coupled to the image intensifier by means of the image distributor. The video camera converts the light from the output screen of the image intensifier into an electrical signal (output video signal) and

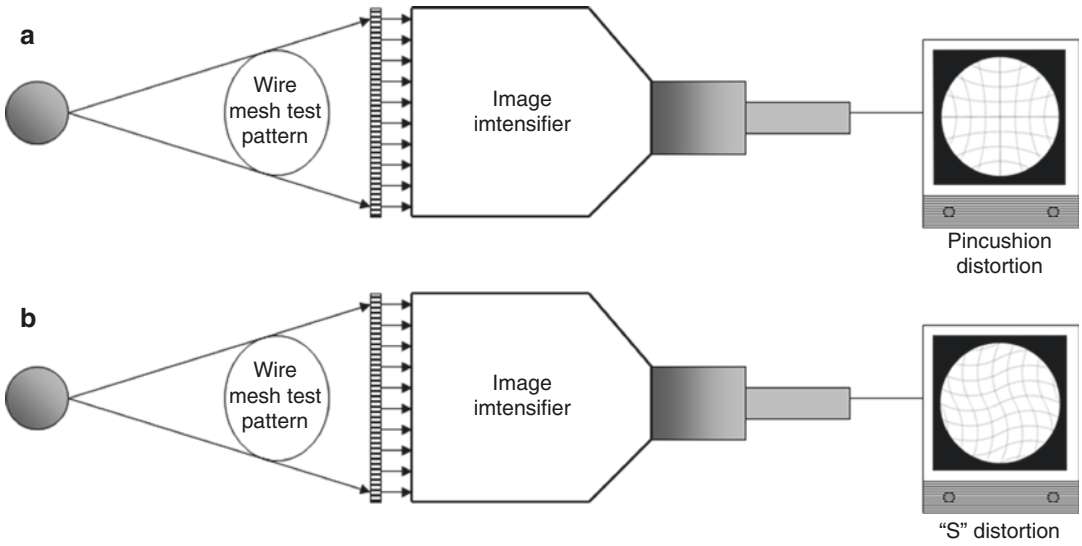

Fig. 6.5 A graphic illustration of two typical image artifacts characteristic of image intensifier fluoroscopy. While pincushion distortion is seen in (**a**), "S" distortion is shown in (**b**). See text for further explanation

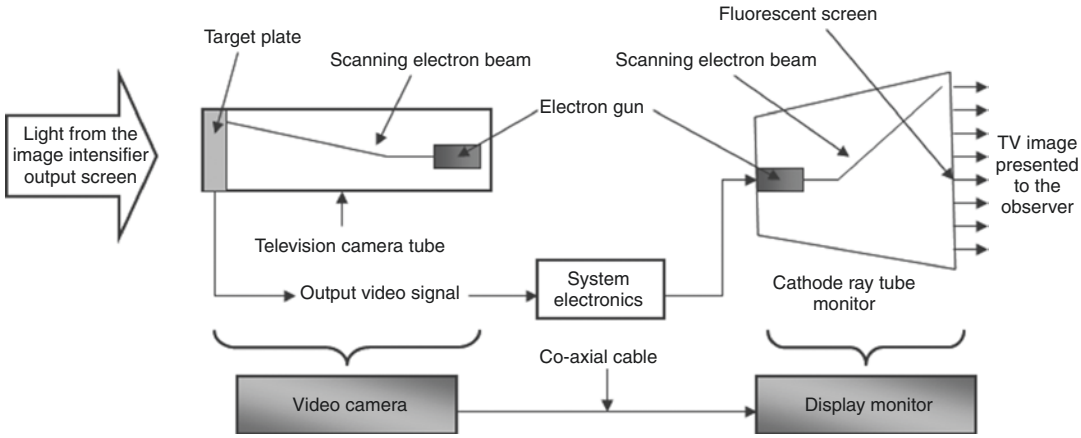

Fig. 6.6 The technical components of a fluoroscopic television chain. The television camera tube is connected to the television monitor by means of a coaxial cable

sent to the television monitor when it is converted into a visible image that can be viewed in real time. Television camera tubes were used in early image intensifier systems, and they were replaced by the CCD camera [1, 2], and therefore they will not be described further in this chapter; however, the interested student should refer to any good radiologic physics or equipment textbook for a description of how these tubes work.

The CCD chip (about 100 mm × 100 mm square) is mounted onto a camera head that is coupled to the output screen of the image intensifier tube. The CCD consists of a matrix (e.g., 1024 × 1024) of pixels that capture the image from the output screen. Each pixel consists of a photosensitive region that produces electrons when struck by light. The electrical charge from each pixel is readout very systematically using suitable electronics to produce an output video

signal that goes to television monitor to create fluoroscopic images displayed in real time.

When compared with television camera tubes, the CCD camera is compact and has a longer life span than television camera tubes. In addition, there is no image lag with the CCD camera, as well as no spatial distortions. Furthermore, the CCD camera has a high dynamic range. As noted by Holmes et al. [7], the dynamic range is a "ratio of the largest detectable signal to the smallest, corresponding to the brightest and darkest regions of the image. The CCD camera has a significant advantage with a dynamic range of 3000:1 compared with the pick-up tube's linear usable range of approximately 1000:1."

The final component in the fluoroscopic television chain is the television monitor. These monitors can be either of the cathode-ray tube (CRT) type or the liquid crystal display (LCD) type. The monitor receives the video signal from the video camera and with suitable electronics uses the signal to create the television image. This image is made up of lines from the scanning process. Scanning can be interlaced in which case 262.5 odd lines (one TV field) are first scanned, followed by 262.5 even lines (one TV field). These two fields are interlaced to create one TV frame that is made up of 525 lines. This type of scanning reduces flickering of the image. At a frequency of 60 Hz (60 cycles/s), 30 frames per second (60 fields/s) can be displayed so that flickering cannot be observed. On the other hand, scanning can be progressive, in which case each line is read sequentially (1, 2, 3, 4, and so on, to 525 lines). Progressive scanning is important in digital fluoroscopy [1].

6.3 Digital Fluoroscopy with Image Intensifiers

The next developmental stage in the evolution of digital fluoroscopy is to digitize the output video signal (either from the television camera tube or the CCD camera) and use a digital computer to process the digitized data and subsequently display the image on a television monitor. To accomplish this task, two new components have been added to the conventional fluoroscopic imaging system. These are the analog-to-digital converter (ADC) and a digital computer. This is now known as an image intensifier-based digital fluoroscopy system, and it consists of an X-ray tube and generator, the image intensifier, the video camera, the ADC, the digital computer, a digital-to-analog converter (DAC), and finally, the television monitor.

6.3.1 X-Ray Tube and Generator

The X-ray tube and generator provide the appropriate radiation beam for imaging in digital fluoroscopy. The X-ray tube is a high-capacity tube and is pulsed in operation. The generator on the other hand is a high-frequency generator and can provide high mA values to be used in digital fluoroscopy, compared with low mA values typical of conventional fluoroscopy. As noted by Bushong [1], the mA is sometimes 100 times higher. These higher values are especially important in digital angiography. The generator ensures that the X-ray beam is pulsed rather than being produced continuously. The pulsing is referred to as "pulsed-progressive fluoroscopy" [1]. Another important point emphasized by Bushong [1] is that "during digital fluoroscopy, the X-ray tube operates in the radiographic mode."

6.3.2 Video Camera

As noted earlier in this chapter, the *video camera* used in fluoroscopy can be either a television camera tube or a CCD camera. While the first-generation systems used television tubes, second-generation systems use CCD cameras. If a television camera tube is used, it must have a higher signal-to-noise ratio (SNR) compared with tubes used in conventional fluoroscopy. While the SNR for conventional fluoroscopy systems is about 200:1, it is 1000:1 for digital fluoroscopy with image intensifiers [1]. Television camera tubes should also have low image lag.

The CCD camera is now used in digital fluoroscopy systems using image intensifiers.

The CCD was already described earlier in this chapter. It is important to note, however, that the CCD camera has extremely high sensitivity and low readout noise level. Images can also be acquired at 60/s compared with 30/s for television tubes.

6.3.3 Analog-to-Digital Converter

The *analog-to-digital converter (ADC)* is an integral component in digital imaging technologies. In digital fluoroscopy, the ADC receives the output video signal (analog signal) from the video camera. Since a digital computer is used, the analog signal must be converted into digital data for processing by the digital computer. The process of digitizing the analog signal requires dividing it (the signal) into a number of parts. This is referred to as sampling (Chap. 2). The unit of the parts is the bit (contraction for binary digit). A bit can be a 1 or it can be a 0. A 2 bit ADC will divide a signal into 4 (2^2) parts. Similarly, a 10 bit ADC will divide the signal into 1024 (2^{10}) parts. The higher the number of bits, the more accurate is the ADC.

6.3.4 Computer System

At the heart of a digital fluoroscopy imaging system is a *digital host computer*. Such a computer is a minicomputer system capable of receiving dynamic digital data from the ADC and processing it quickly for image display and subsequent storage. It is not within the scope of this chapter to describe the details of the computer system; however, the following points are noteworthy:

- The computer operates on the data it receives from the ADC in a matrix format. This means that the digital image is a matrix of pixels, and for digital fluoroscopy, matrix sizes of 512 × 512 and 1024 × 1024 are typical.
- Each pixel in the image contains the atomic number and mass density characteristic of the tissue, and a single number for the pixel represents this information.
- The matrix of numbers is transformed into a grayscale image as illustrated in Fig. 6.7. The image can be described by the term "bit depth," what describes the number of shades of gray that a single pixel in the matrix (image) can assume. For example, while in a 3 bit depth image, each pixel can have 8 (2^3) shades of gray, an 8 bit depth image will provide 256 (2^8) shades of gray for each pixel.
- The spatial resolution for a digital fluoroscopic image depends on the pixel size. As the matrix size increases for the same field of view (FOV), the pixel size decreases, and the image appears sharper.
- Images displayed on the television monitor can be post-processed using a number of different image processing operations. These include last image hold, grayscale image manipulation, and edge enhancement, for

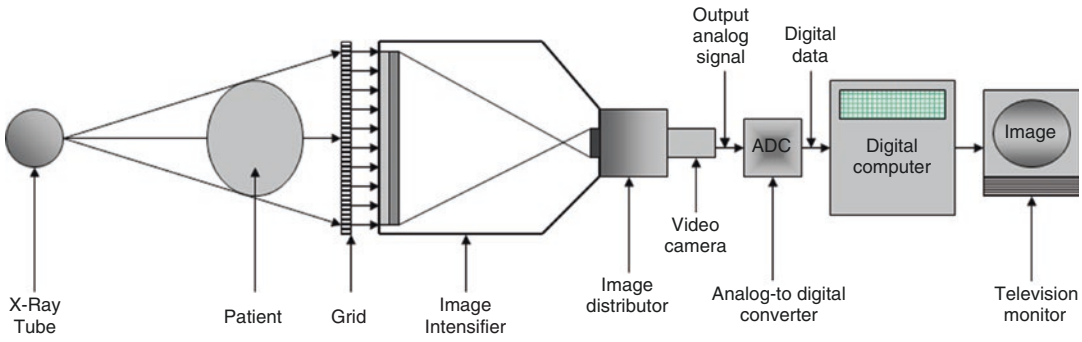

Fig. 6.7 The addition of the analog-to-digital converter (ADC) and a digital computer, which are two critical pieces of equipment of an image intensifier-based digital fluoroscopy imaging system

example. Such processing is a function of the digital computer, and it is intended to alter the image in several ways to enhance diagnostic interpretation.

6.4 Digital Fluoroscopy with Flat-Panel Detectors

Recently, digital fluoroscopy systems with the image intensifier and video camera in the imaging chain have been replaced with *digital fluoroscopy systems using flat-panel detectors (FPDs)*. Specifically, FPDs have replaced image intensifiers and video cameras (television camera tubes and CCD cameras). Digital fluoroscopy with FPDs have become commonplace in angiography, and only a few systems are currently being used for gastrointestinal tract fluoroscopy (at the time of writing this chapter). This section will therefore focus on the fundamental principles guiding the performance of FPDs in digital fluoroscopy.

6.4.1 Limitations of Image Intensifier Technology

The image intensifier tube technology poses several problems when used for fluoroscopic imaging. These problems were identified earlier in this chapter and include veiling glare, vignetting, image lag, and pincushion and "S" distortions. Additionally, light and electron scattering within the tube degrade the image contrast (veiling glare), and image magnification results in increased dose to the patient.

6.4.2 Equipment Configuration

The overall equipment configuration of a FPD digital fluoroscopy imaging system (real-time imaging) is shown in Fig. 6.8. It is clear that the most significant difference between this system and the image intensifier-based system shown in Fig. 6.7 is the presence of the FPD. FPDs used in radiographic imaging (Chap. 4) produce static images and are therefore referred to as static FPDs. One significant and important technical characteristic of an FPD for fluoroscopy is that it must be capable of producing dynamic images that can be displayed and viewed in real time. For this reason, these detectors are sometimes referred to as *dynamic FPDs*.

The components shown in Fig. 6.8 include the X-ray tube, the patient, the grid, the dynamic FPD, the host computer, and the television display monitor. The next subsection of this chapter will only highlight the basics of the dynamic FPD.

6.4.3 Types of Dynamic FPDs

Two types of dynamic FPDs are currently available for digital fluoroscopy, namely, the CsI a-Si TFT indirect digital detector and the a-Se TFT direct digital detector. In general these detectors are similar in design to the static FPDs used for

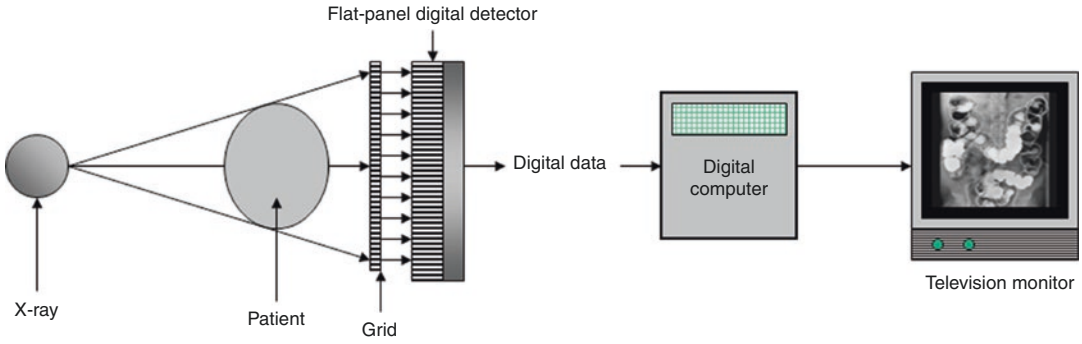

Fig. 6.8 The overall equipment of a digital fluoroscopy imaging system using flat-panel digital detector

radiographic imaging described in Chap. 4. There are a few significant differences, however, and they will be reviewed subsequently.

Various medical imaging vendors utilize different dynamic FPDs in their digital fluoroscopy systems, and the interested reader should refer to the various manufacturers' websites for detailed specifications of these systems.

6.4.4 Characteristics of Dynamic FPDs

The characteristics that are worthy of consideration in this chapter are the dimensions of the detector, matrix sizes, pixel considerations, and the zoom feature. Typical dimensions of these detectors vary; however, 31 cm × 31 cm, 35 cm × 35 cm,

30 cm × 40 cm, and 41 cm × 41 cm are commonplace. Recently, 43 cm × 43 cm dynamic FPD has become available for both digital fluoroscopy and digital radiography.

Matrix sizes vary as well depending on the physical dimensions of the detector. Generally, larger detectors have larger matrix sizes, and typical sizes include 1024 × 1024, 2304 × 2304, and 2048 × 2048.

For digital fluoroscopy detectors, the *pixel size* is larger than the pixel size used in digital radiography detectors. In some systems it is possible "to adjust the pixel size by binning four pixels into one larger pixel. Such dual-use systems have pixels small enough to be adequate for radiography (e.g., 100–150 μm), but the pixels can be binned to provide a detector useful for fluoroscopy (e.g., 200–300 μm)" [2].

Another characteristic feature of the dynamic FPD is that they offer "zoom" modes. In this respect, the operator can zoom into the survey image and examine the details of smaller structures.

6.4.5 Operating Principles

There are other significant differences between FPDs for digital radiography and dynamic real-time fluoroscopy, most notably the nature of the readout electronics. Such complexity is beyond the scope of this book; however, the interested reader should refer to a paper by Lai et al. [8]. One primary consideration is that the electronics must facilitate high *frame rates* and fast data transfer rates. Frame rates of 15–30 fps or greater are possible at readout speeds of 30–50 ms [5, 7]. Additionally, dynamic FPDs can operate in at least two readout modes: the frame rate (fps) in continuous X-ray mode and the fps in the pulsed X-ray mode.

In general, the operational elements of a dynamic FPD involve three sequences that result in a single image that must be completed in at least 33 ms for fluoroscopy. These elements include initialization, integration, and readout [7]. While initialization prepares the detector electronics for X-ray exposure, integration and readout are intended to collect the detector signal (analog signal) for subsequent digitization and image display.

6.4.5.1 Magnification

Magnification using image intensifier fluoroscopy was described above. In summary magnification provides increased spatial resolution but at the expense of increased dose to the patient. In general the increase is about 2.2 times that is used in the full-field mode of intensified fluoroscopic operation. With flat-panel digital fluoroscopy systems, there are two methods of magnification: (a) electronic magnification (zoom) and (b) binning. Binning is illustrated in Fig. 6.9a. With electronic magnification, there is no increase in spatial resolution, and both original and magnified images have the same signal-to-noise ratio (SNR). Additionally, "binning has the disadvantage of less spatial resolution because the effective area of each image pixel is four times larger, and it has the advantage of lower data rates and less image mottle than ungrouped pixels" [3].

In Fig. 6.9b, two field of views (FOVs) are shown, a central smaller FOV and a larger FOV. As explained by Nickoloff [3], "for smaller FOVs, collimation is used to select only the central portion of the FPD for imaging, which is similar to the process used with image intensifier

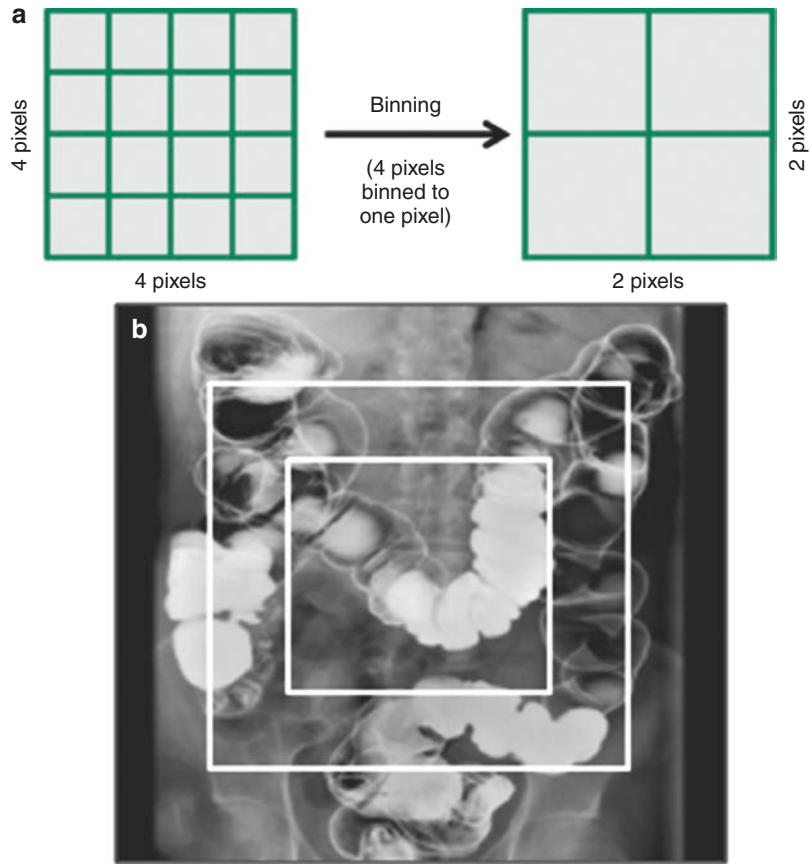

Fig. 6.9 Electronic magnification in digital fluoroscopy includes binning (**a**) and electronic zooming (**b**). See text for further explanation

fluoroscopy systems; thus, information from a smaller anatomic area is spread across the display monitor or magnified (Fig. 6.9b). When smaller FOVs are used, the data rate is lower, and binning is no longer required. Unlike image intensifier fluoroscopy systems, the spatial resolution of FPD fluoroscopy systems is the same for all FOVs—if no binning is employed. For those larger FOVs when binning is employed, the spatial resolution dramatically decreases to 50% of the value without binning. For FPD systems, there is a dramatic, discrete step change in spatial resolution between small and large FOVs."

6.4.6 Advantages

Compared with image intensifier-based digital fluoroscopy, dynamic FPD fluoroscopy offers several advantages. These include distortion-free images, improved contrast resolution, high detective quantum efficiency (DQE), and uniform image quality over the whole displayed rectangular image (a format that makes efficient use of the rectangular display of television monitors), while it is circular for intensifier-based systems as illustrated in Fig. 6.10a and b, respectively [1].

Dynamic FPDs also make use of scattered radiation grids that can be removed when imaging children. The use of grids during the examination will increase the dose to the patient. It is vital to note that during its use, the grid lines are oriented diagonally to the detector matrix in an effort to eliminate any aliasing artifacts [4]. Finally, digital fluoroscopy using FPDs is much more compact in design, and this feature enables excellent access to the patient during the examination.

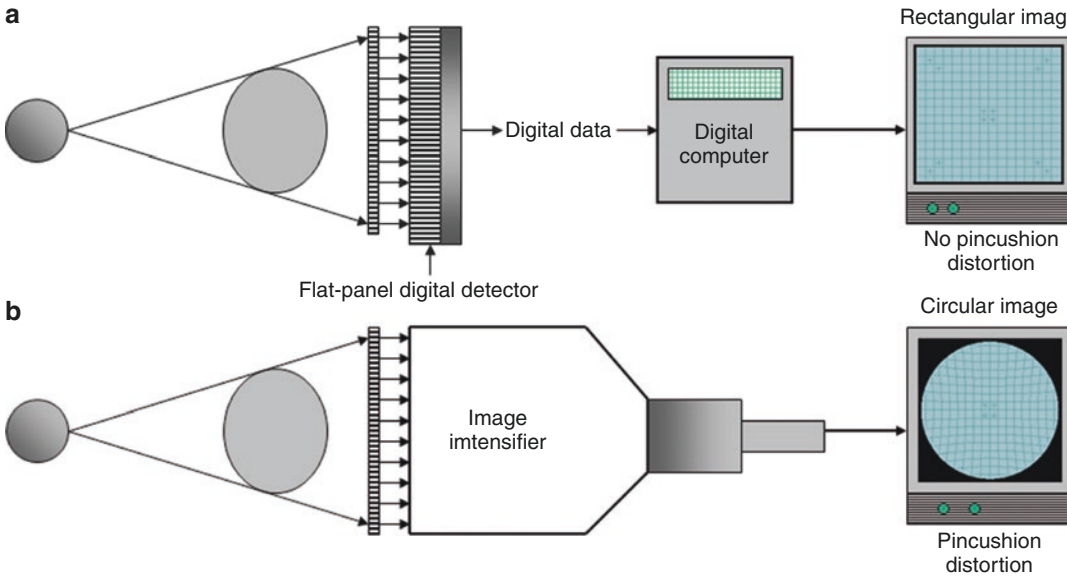

Fig. 6.10 The image display format when using a flat-panel digital fluoroscopy imaging system is rectangular (**a**), while it is circular for an image intensifier-based fluoroscopy system (**b**). The rectangular format matches the display format of the television monitor. Note that the pixels in **b** are distorted especially at the periphery of the image, while the pixels in the **a** are not distorted

6.4.7 Connectivity

Digital fluoroscopy imaging systems can be configured with several DICOM standards to ensure integration with Picture Archiving and Communication Systems (PACS) and information systems such as the radiology information system (RIS) and the hospital information system (HIS). For example, the DICOM "query/retrieve" service classes can provide direct access to archived images (in PACS) for display on the radiologist workstation.

6.5 Digital Image Post-processing

All digital imaging modalities including digital fluoroscopy make use of various types of image post-processing algorithms (software) essentially to manipulate the image presented to the observer. The purpose of this operation is essentially to enhance diagnostic interpretation [9]. Since digital fluoroscopy can be applied to gastrointestinal (GI) tract imaging and digital subtraction angiography (DSA), image post-processing operations are specific to each application. For example, while grayscale image manipulation is common to GI fluoroscopy, road mapping is an operation commonly used in DSA. DSA will be described briefly later in this chapter.

Image post-processing operations specifically for digital fluoroscopy include grayscale image manipulation, temporal frame averaging, last image hold, and edge enhancement. Each of these will be described briefly.

6.5.1 Grayscale Image Manipulation

Grayscale image manipulation was described in detail in Chap. 2. In review, the purpose of grayscale image manipulation is to change the contrast and brightness of an image displayed on the monitor in order to facilitate diagnostic interpretation [9].

Recall that a digital image is made up of a matrix of pixels where each pixel is assigned a

number. Each number on the other hand corresponds to a gray shade. An 8 bit image, for example, will consist of 256 (2^8) ranging from 0 to 255. This image will also consist of 256 shades of gray, where low numbers are dark and higher numbers are bright. The range of the numbers is defined as the window width (WW), and the center of the range is defined as the window level (WL). While the WW changes the image contrast, the WL changes the brightness of the image. Fig. 6.11 clearly demonstrates the effect of the WW (a) and the WL (b) on image contrast and brightness, respectively.

6.5.2 Last Image Hold

Last image hold (LIH) is an image processing technique used to reduce the radiation dose to the patient. When the X-ray beam is turned on, images are obtained. The LIH operation displays the last frame continuously when the X-ray beam

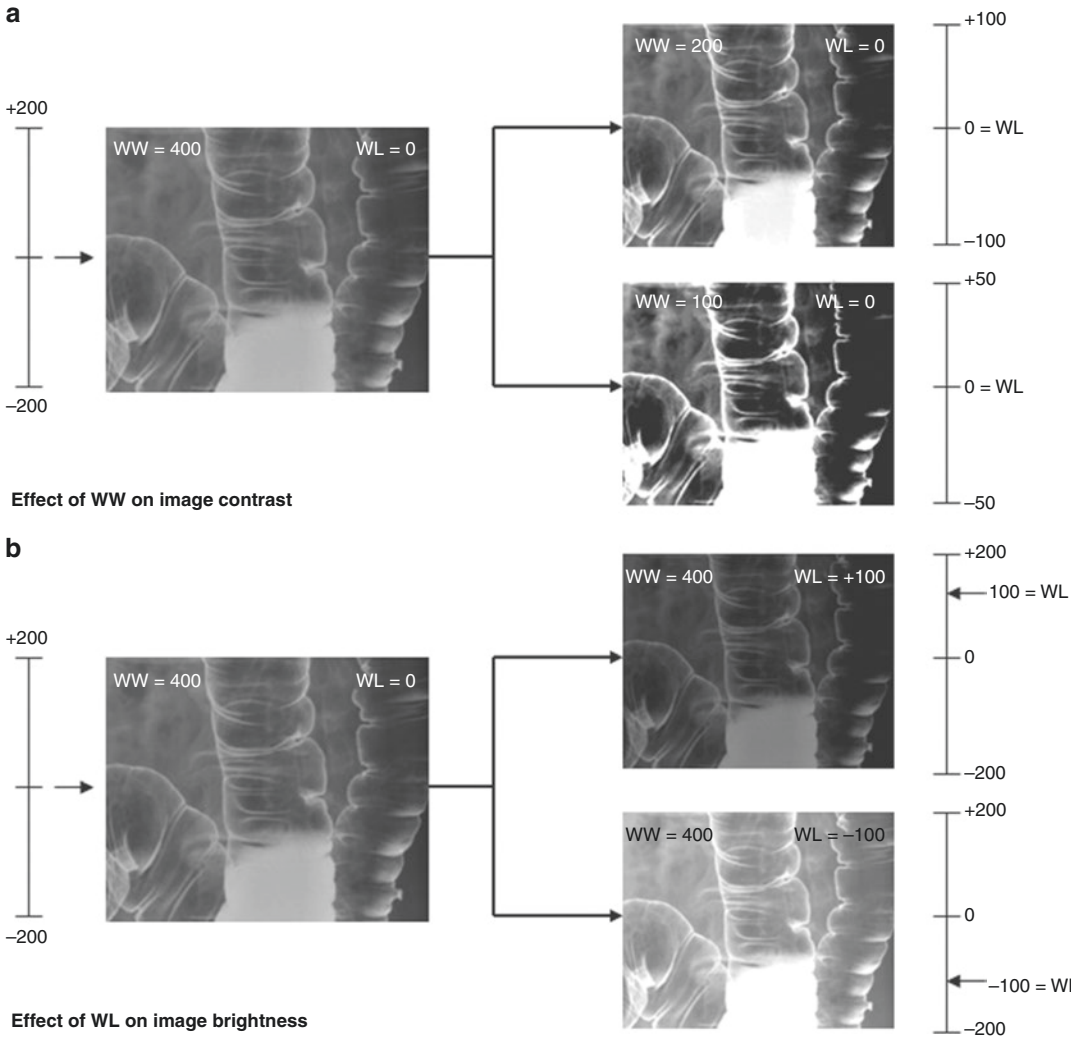

Fig. 6.11 The effect of changing the window width (WW) and window level (WL) settings on the grayscale appearance of the image. See text for further explanation

is turned off. The process repeats itself when the beam is turned on once again.

6.5.3 Temporal Frame Averaging

The purpose of *temporal frame averaging* is to reduce the noise present in an image "by continuously displaying an image that is created by averaging the current frame with one or more previous frames of digital fluoroscopic image data" [10]. For example, averaging five frames will reduce the noise by about 44%; however, as more and more frames are averaged, image lag results.

6.5.4 Edge Enhancement

Edge enhancement is an image-sharpening post-processing operation (Chap. 2). Several image-sharpening algorithms are currently available, and these have been described by Seeram and Seeram [9].

6.5.5 Proprietary Post-processing Techniques

It is clearly understandable that different vendors will have post-processing algorithms specific to their equipment, and students and technologists alike will experience this when they work with different vendor equipment. In addition, one can explore appropriate websites for detailed specifications and illustrations.

6.6 Digital Subtraction Angiography: A Brief Overview

As noted earlier in the chapter, dynamic FPD detectors were first used in angiography rather than in GI tract fluoroscopy. Dynamic FPDs have replaced image intensifier/video camera technology in angiographic imaging. Since contrast material is used in angiography, and precontrast images are digitally subtracted from postcontrast images during the procedure, the imaging process is referred to a *digital subtraction angiography* (DSA). DSA is beyond the scope of this book; however, the information presented below is noteworthy. There are two methods of DSA: temporal subtraction and energy subtraction.

6.6.1 Temporal Subtraction

Temporal subtraction involves the digital subtraction of images in time. In general, a precontrast image referred to as a mask image is first obtained. Subsequently, postcontrast images are digitally subtracted from the mask image. Six postcontrast images (frames) are used in the subtraction sequence to provide images of only the contrast-filled vessels. All other overlying anatomical structures are removed [1].

6.6.2 Energy Subtraction

The *energy subtraction* operation is based on subtraction of images taken at different kVs. Images are recorded based on subtraction of energies slightly above and slightly below the k-absorption edge of the contrast material used for the examination. For a further description of temporal and energy subtraction techniques, the student should refer to Bushong [1].

References

1. Bushong S. Radiologic science for technologists. 11th ed. St Louis: Elsevier-Mosby; 2017.
2. Bushberg JT, Seibert JA, Leidholdt EM Jr, Boone JM. The essential physics of medical imaging. 3rd ed. Philadelphia: Lippincott Williams and Wilkins; 2012.
3. Nickoloff EL. Survey of modern fluoroscopy imaging: flat-panel detectors versus image intensifiers and more. Radiographics. 2011;31:591–602.
4. Seeram E, Brennan P. Radiation protection in diagnostic X-ray imaging. Burlington, MA: James & Bartlett Learning; 2017.
5. Jones AK, Balter S, Rauch P, Wagner LK. Medical imaging using ionizing radiation: optimization of dose and image quality in fluoroscopy. Med Phys. 2014;41(1):014301.

6. Wang J, Blackburn TJ. X-ray image intensifiers for fluoroscopy. Radiographics. 2000;20:1471–7.
7. Holmes DR, Laskey WK, et al. Flat-panel detectors in the cardiac catheterization laboratory: revolution or evolution-what are the issues? Catheter Cardiovasc Interv. 2004;63:324–30.
8. Lai J, Safavian N, et al. Active pixel TFT arrays for digital fluoroscopy in a-Si:H technology. Mater Res Soc Symp Proc. 2005;862:635–40.
9. Seeram E, Seeram D. Image postprocessing in digital radiology: a primer for technologists. J Med Imaging Radiat Sci. 2008;39(1):23–43.
10. Pooley RA, McKinney MI, Miller DA. Digital fluoroscopy. Radiographics. 2001;21:521–34.

Full-Field Digital Mammography

7

Abstract

Mammography is defined as radiography of the breast. Digital mammography (DM) or full-field digital mammography (FFDM) has become commonplace in medical imaging departments. FFDM overcomes the limitations of film-screen mammography (FSM). Two major limitations include a limited dynamic range, and that the display characteristics such as brightness and contrast are fixed due to the chemical processing of the film. FFDM is radiography of the breast using a digital detector coupled to a digital computer that makes use of digital image processing techniques to enhance the visibility of detail and contrast of the image, in an effort to improve the detectability of breast lesions. FFDM consists of five steps which includes data acquisition, analog-to-digital conversion (ADC), digital image processing, image display, image storage, archiving, and communications via the picture archiving and communications system (PACS). Four types of digital detector systems are used for FFDM: flat-panel phosphor system, charge-coupled device (CCD) system, Flat-Panel amorphous Selenium (a-Se) System, and a computed radiography (CR) FFDM system. Detectors for FFDM must be capable of providing a spatial resolution of at least 10 line pairs/mm (lp/mm) to improve lesion detectability. Another performance characteristic of a FFDM detector is the detective quantum efficiency (DQE), which provides an indication of how well the FFDM imaging system can efficiently transfer the input signal-to-noise ratio (SNR) (at the detector) to the output SNR (image displayed on the monitor) so that it is useful to the observer in making a diagnosis. Digital image processing is an essential feature of FFDM and include operations such as windowing, measurement and annotation tools, as well as various sophisticated digital post-processing techniques, such as frequency processing for enhancing the sharpness of an image as well as manual intensity windowing (MIW), histogram-based intensity windowing (HIW), mixture-model intensity windowing (HMIW), contrast-limited adaptive histogram equalization (CLAHE), unsharp masking, and peripheral equalization. Applications of FFDM include computer-aided detection and diagnosis, digital breast tomosynthesis (DBT), and contrast-enhanced digital mammography. DBT is also referred to as three-dimensional (3D) mammography and is a relatively new technique which has increased attention in the literature.

E. Seeram, *Digital Radiography*, https://doi.org/10.1007/978-981-13-3244-9_7

7.1 Introduction

Mammography is defined as radiography of the breast, and it is a prime example of soft tissue radiography, since the breast is composed of soft tissues such as adipose (fat), fibrous, and glandular tissues. Mammography was developed as a dedicated imaging technique to detect breast cancer that has become prevalent among women in North America. The imaging technique is dedicated because it uses equipment designed especially to produce an X-ray beam that will allow for maximum X-ray absorption by the soft tissues, microcalcifications, and thin fibers so that they can be shown on radiographic film with excellent spatial resolution or detail. Mammography must be able to show the contrast between a lesion that is located in the breast and the normal anatomy that is around that lesion. This imaging technique is referred to as screen-film mammography.

It is important to note that sections of this chapter have been previously published from my PhD thesis, entitled, *Optimization of the Exposure Indicator as a Dose Management Strategy in Computed Radiography.* PhD Dissertation (Charles Sturt University, New South Wales, Australia, 2014).

7.2 Screen-Film Mammography: A Review of the Basics

It is not within the scope of this chapter to describe the physics and technical details of screen-film mammography (SFM), and therefore the interested reader should refer to any good Radiologic Physics textbooks [1, 2] for a comprehensive coverage of these topics.

7.2.1 The Imaging Process

The basic process for *screen-film mammography* (SFM) is shown in Fig. 7.1. A dedicated X-ray tube produces a special X-ray beam (soft X-rays) that passes through the breast to fall upon the mammography cassette. This cassette contains the intensifying screen and the film that has been made for use in mammographic imaging. X-rays cause the screen to emit light, and subsequently

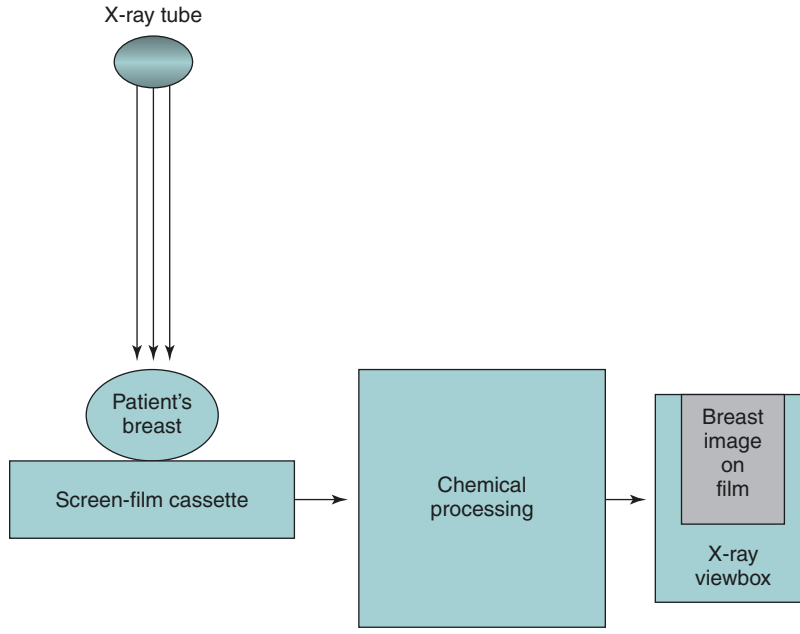

Fig. 7.1 The major components of a screen-film mammography imaging system. The radiation passes through the patient's breast to create a latent image on the film. The film is processed using chemical solutions to render the latent image visible. The image in then displayed on a light box (viewbox) for viewing and interpretation by a radiologist

a latent image is formed on the film. The film is processed using a chemical processor, to render the latent image visible. This film image is then displayed on a light viewbox for interpretation by a radiologist who makes a diagnosis of the patient's medical condition. In this entire process, the film acts as the acquisition device, the display medium, and subsequently the storage medium, since it is placed in an envelope and housed in a room for long-term storage and archiving.

SFM has been successful for decades since it offers several advantages [3] most notably; it provides not only high image contrast but also high spatial resolution of about 15–20 line pairs/mm limiting resolution which is needed to detect specks of calcium hydroxyapatite (microcalcifications) with diameters of around 0.01 mm (100 μm) [1].

7.2.2 Limitations of SFM

Apart from these advantages, SFM poses several limitations such as a limited dynamic range (narrow exposure latitude) of the film as described in Chaps. 1, 3, and 4. The film will only respond to a narrow range of exposures, and therefore the technologist must be extremely careful in selection of the optimum exposure factors to provide the best image contrast. Such contrast falls on the slope of the characteristic curve (the H-D = Hurter-Driffield curve). Additionally, the display characteristics of film such as its brightness and contrast are fixed once the film is developed in the chemical processor. If the radiologist needs a lighter or darker image or if the image contrast needs to be changed, then the technologist must perform the exam using a different set exposure technique factors to demonstrate these required characteristics. Another limitation of FSM is that the film serves three roles: acquisition, display, and storage as mentioned above.

Over the years SFM progressed with significant developments such as technical and clinical improvements and regulatory approval. One such significant technical development/improvement is digital mammography designed to overcome the limitations of SFM.

7.3 What Is Full-Field Digital Mammography?

Full-field digital mammography (FFDM) is radiography of the breast using a digital detector coupled to a digital computer that makes use of digital image processing techniques to enhance the visibility of detail and contrast of the image, in an effort to improve the detectability of breast lesions [4]. While the overall equipment for positioning the patient for the examination and the digital control workstation is shown in Fig. 7.2a, the major components of a FFDM imaging system and imaging steps are clearly illustrated in Fig. 7.2b.

In FFDM, a digital detector replaces the film-screen image receptor used in conventional film-screen mammography. X-rays passing through the breast fall upon the digital detector to produce a latent image that is subsequently processed by a digital computer. The digital image is displayed on a workstation monitor for viewing and interpretation by a radiologist. In the digital system, the image acquisition, storage, and display are performed in separate components. Additionally, the digital image can be manipulated further to suit the viewing needs of the observer and removes the problems of overexposure and underexposure.

The development of FFDM dates back more than a decade ago. In September 1991, the National Cancer Institute was advised by a group of well-known breast imaging experts to develop the technique of digital mammography [5]. Today, several research groups and manufacturers are actively engaged in developing FFDM into a useful clinical imaging tool. As of 2017, examples of major manufacturers who are actively engaged in FFDM development include Hologic (Selenia Dimensions Avia 3000; Selenia Dimensions system 6000 and 9000), Fischer Imaging (MammoCAT™), Fuji Medical Systems (ASPIRE Cristalle FFDM System), General Electric Healthcare (Senographe Pristina™

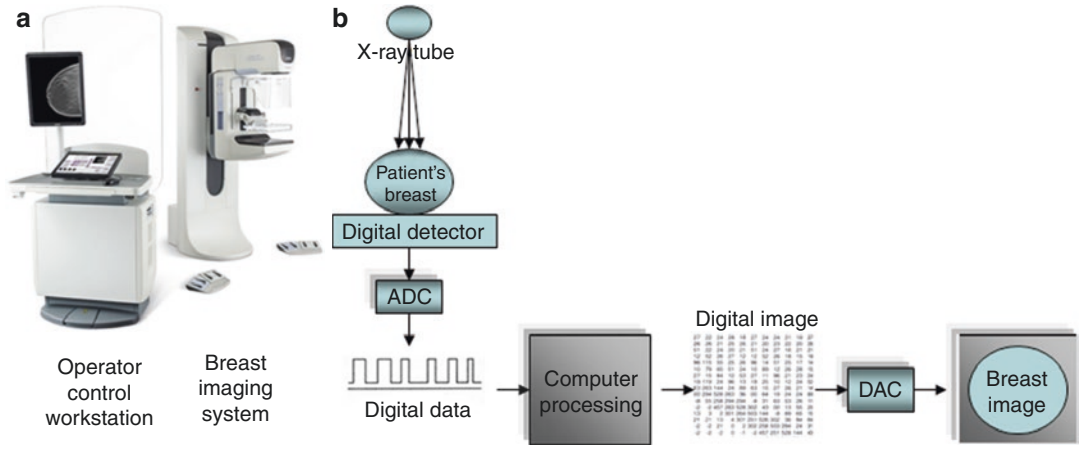

Fig. 7.2 The major components of a FFDM imaging system. While the equipment for positioning the patient for the examination and the digital control workstation is shown in (**a**), the basic steps in producing the image are clearly illustrated in (**b**) (Photograph Courtesy of Hologic)

System), and Siemens Healthineers {MAMMOMAT Revelation (awaiting FDA clearance as of December 2017)}. These manufacturers have developed several digital mammography imaging systems, each based on a different set of technical principles. The interested reader is encouraged to explore these products on the websites of these vendors.

7.3.1 Advantages of FFDM

FFDM uses a digital detector and computer processing to generate images, as opposed to chemical processing of film characteristic of SFM. Because of this, FFDM offers the following advantages over SFM [4, 5]:

- The digital detector offers a wider dynamic range. While SFM offers a dynamic range of about 40:1, it is 1000:1 for FFDM.
- Greater contrast resolution especially for dense breast tissue.
- Use of digital image post-processing operations to enhance image quality.
- The ability to communicate with a picture archiving and communication system (PACS).
- Computer-aided detection (CAD) and diagnosis.
- Digital breast tomosynthesis (DBT).
- Contrast-enhanced DM.

Digital image post-processing, CAD, telemammography and digital tomosynthesis, and contrast-enhanced DM will be described briefly later.

7.4 Technical Requirements for FFDM

FFDM involves at least five fundamental steps as shown in Fig. 7.3. These include data acquisition, analog-to-digital conversion (ADC), digital image processing, image display, image storage, archiving, and communications via the PACS. Once in the PACS, images and text data can be sent to remote locations via communication networks.

7.4.1 Data Acquisition

Data acquisition is the first step in producing a FFDM image. The data acquisition consists of the X-ray tube and generator systems coupled to the digital detector imaging system. X-rays pass through the breast and fall upon the digital

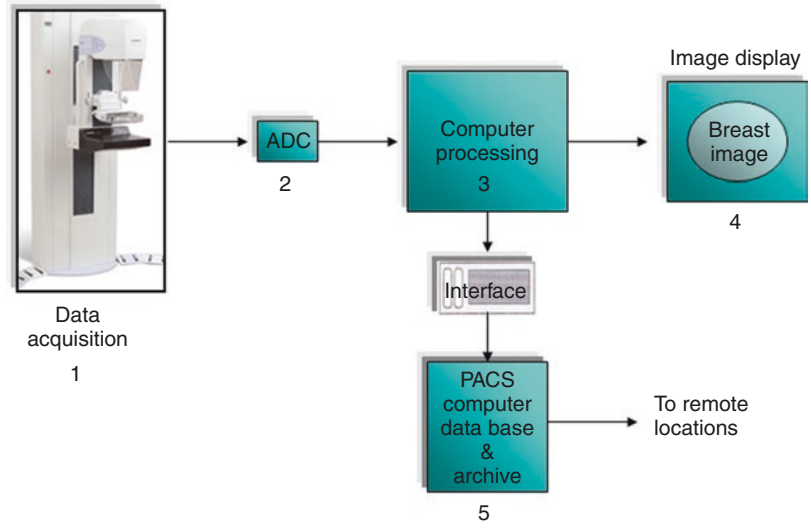

Fig. 7.3 Digital mammography involves at least five fundamental steps: data acquisition, analog-to-digital conversion (ADC), digital image processing, image display, image storage, archiving, and communications via the PACS (Photograph Courtesy of Hologic)

detector to create an electronic signal that must subsequently be digitized for input into a digital computer. Currently, there at four types of digital detector systems used for FFDM: flat-panel phosphor system, charge-coupled device (CCD) system, Flat-Panel amorphous Selenium (a-Se) System, and a computed radiography (CR) FFDM system [1, 4, 5]. Each of these will be described briefly later in this chapter.

Detectors for FFDM must be capable of providing a high spatial resolution. While FS can show about 20 line pairs/mm (lp/mm), FFDM must be capable of at least 10 lp/mm to improve lesion detectability [1] and must have an image area of at least 24 cm × 30 cm in order to capture the entire breast. Additionally since the pixel size affects the spatial resolution, pixels must be spaced about 25 μm (0.025 mm) apart. For a large detector, 9600 × 12,000 matrix size is a fundamental requirement [6]. In addition, as noted by James [6] "for good image quality, a high intensity signal is required with low-intensity noise-a good Signal-to-Noise Ratio (SNR)." Recall that the SNR is describes the information quality in the image.

Another measure of the detector's performance in FFDM is the detective quantum efficiency (DQE). This is how well the FFDM imaging system can efficiently transfer the input SNR (at the detector) to the output SNR (image displayed on the monitor) so that it is useful to the observer in making a diagnosis. The DQE is expressed as follows:

$$DQE = SNR_{out}^{2} / SNR_{in}^{2}$$

In a perfect imaging system, the DQE would be equal to one (100%). The DQE for F-S mammography system is about 45% [7]. The reader should refer to the various manufacturers for the DQE for their FFDM systems.

7.4.2 Analog-to-Digital Conversion

The digital detectors convert X-ray photons to an electronic signal (analog signal) that must first be digitized for input into a digital computer. FFDM systems required *ADC*s with very high digitization capability. FFDM systems may use 12–14 bit ADCs. A 14 bit ADC can divide an electronic signal into 16, 384 (2^{14}) discrete parts. This is rated as excellent digitization and will enable images to be displayed with excellent grayscale bit resolution [6].

7.4.3 Digital Image Processing

The digital computer is central to a digital mammography imaging system. The computer

generates a digital mammographic image, using the digital data it receives from the ADC and using a defined set of *image processing algorithms*. It is not within the scope of this chapter to describe these algorithms. In addition, a number of post-processing operations are available to enhance images to suit the viewing needs of the observer. These operations, for example, include windowing, measurement and annotation tools, as well as various sophisticated digital post-processing techniques, such as frequency processing for enhancing the sharpness of an image. These will be discussed later in the chapter.

7.4.4 Image Display

The digital output from the computer must be converted back into an analog signal so that it can be displayed on a monitor workstation (soft copy image) that has a display resolution of at least 5 megapixels [4]. Image display also takes into consideration the luminance and contrast resolution of the display monitors.

Two types of display technologies for FFDM are in use today, the cathode ray tube (CRT) and the liquid crystal display (LCD). In the past, Samei [8] has described the technological and psychophysical factors for these types of displays. The psychophysical factors that play a role in image interpretation include contrast resolution, and noise. It is interesting to note that "optimal display of mammograms is achieved by taking these factors into consideration and by using time-efficient, intuitive, and reader-specific user interfaces" [8].

FFDM computer display monitors should also be located in rooms that have ambient light of less than 5 lux[2]. Various studies have evaluated the performance of radiologists on soft-copy images, and while it is not within the scope of this chapter to describe the details of image display, the results of these studies indicate that "the lower luminance and contrast resolution of soft copy systems, compared with those printed on film displays, do not significantly affect the interpretation performance of radiologists" [8].

7.4.5 PACS Integration

PACS are now commonplace in hospitals and the next consideration for FFDM is to send all the digitally acquired images to the PACS. The PACS is characterized by several components working together to achieve a common goal. These components include image storage, image archiving and image communications. Of course the FFDM acquisition components are connected to the PACS, as shown in Fig. 7.4. The

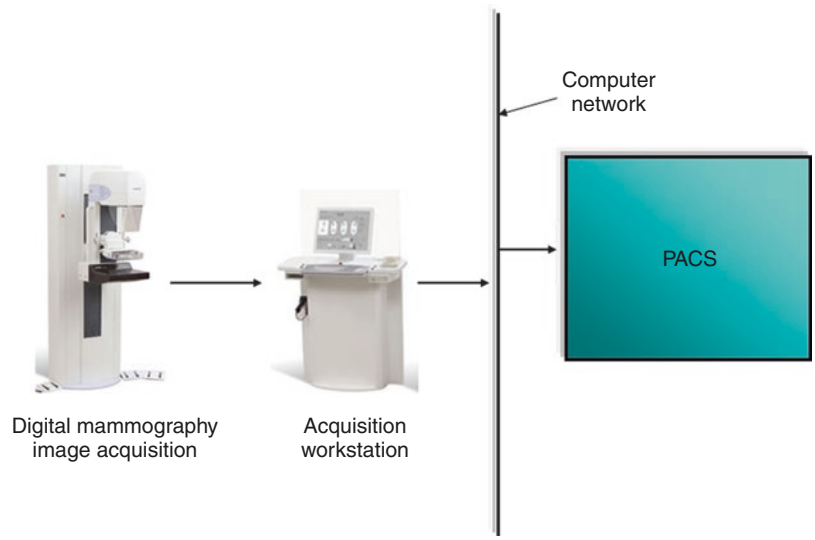

Fig. 7.4 The digital mammography image acquisition system and the acquisition workstation are connected to the PACS via computer networks and an interface (not shown) that facilitates the communication of images to the PACS (Photographs Courtesy of Hologic)

purpose of the PACS is to improve the management efficiency of the large amount of FFDM images, including storage, retrieval, and communication of images.

7.5 Types of Digital Detector Systems for FFDM

As noted earlier in the chapter, there are four different types of FFDM systems available commercially based on the digital detector structure and function. These include type 1 digital detectors such as the Flat-Panel Scintillator/amorphous Silicon (a-Si) System, type 2 digital detectors such as the charge-coupled device (CCD) system, type 3 digital detectors such as the computed radiography (CR) FFDM system, and type 4 digital detectors such as the Flat-Panel amorphous Selenium (a-Se) System. These systems are available from specific mammography vendors.

The overall basic principles of these detectors were described in Chap. 1 and subsequently elaborated in Chaps. 3 and 4. In review, the following brief principles of how each works are noteworthy:

7.5.1 Flat-Panel Scintillator/ Amorphous Silicon (a-Si) FFDM System

The basic structure of the *flat-panel a-Si FFDM detector* system consists of three major components: a scintillator, a thin film transistor (TFT) array, and a glass support. This detector system is sometimes referred to as an indirect conversion system, only because the detector first converts the X-ray photons falling upon it into light photons. Subsequently the light is converted into electricity by the TFT.

The scintillator is made up of cesium iodide (CsI) activated by thallium. The CsI phosphor is deposited on top of the a-Si TFT array in a needle-like fashion (to reduce lateral dispersion of light in an effort to improve image detail) and converts X-rays into light. The light falls upon the a-Si TFT array that is constructed as a matrix of photodiodes and is subsequently converted into electrical signals. These signals are sent to a digitizer and are converted into digital data for input into a digital computer. The computer performs the appropriate digital processing to produce a digital image of the breast.

7.5.2 Charge-Couple Device (CCD) FFDM System

The *CCD FFDM detector* is a type 2 detector. The main structural components include a thallium-activated CsI scintillator phosphor similar to the one described above, deposited upon a matrix of CCDs. This system is also referred to as an indirect conversion system simply because X-rays are first converted into light and not directly into electricity. In addition, the CsI phosphor is coupled to the CCD array via a fibreoptical system that directs the light from the CsI to the CCD array.

The CCD array is arranged as a matrix of pixels that are sensitive to the light from the CsI phosphor. The CCD system converts the light into electrical signals that are read out in a very systematic way and sent to the digitizer that generates digital data for input into a digital computer for processing. The result of computer processing is a digital image of the breast.

This system is somewhat different from the system described above. A fan beam from the X-ray tube falls upon the detector that "has a long, narrow, rectangular shape, with dimensions of approximately 1 × 24 cm. The X-ray beam is collimated into a narrow slot to match this format" [7]. For this reason this system is also referred to as slot-scanning FFDM system, and it "has a distinct advantage over the area detectors in that it is very compact and therefore the detector assembly costs less. Also the system has excellent scatter rejection due to the small breast volume exposed at any time. The need for a grid does not arise in these systems, thereby reducing the overall breast dose. However, these systems need powerful X-ray tubes and generators and elaborate signal readout and image reconstruction. Also relative to other mammography

systems, the system requires longer breast compression" [4].

7.5.3 Computed Radiography (CR) FFDM System

CR was introduced around1981 by Fuji Medical Systems and has found widespread applications in general radiographic imaging. Fuji however has extended its use to mammographic imaging.

The basis for CR is *photostimulable luminescence*. This means that when *photostimulable phosphors* such as barium flurohalide (e.g., barium fluorobromide) are exposed to X-rays, electrons are raised to higher energy state and are stored as a latent image. To render this latent image visible, the phosphor is exposed to a laser light, and the electrons return to their original state. In doing so, light is emitted and subsequently converted into electrical signals that are digitized for computer processing, to produce a CR image. The phosphor is deposited on a special support called the imaging plate (IP) that can be placed in a cassette and looks similar to a film-screen cassette; however, the IP replaces the film, and it can be used again and again for several hundred exposures.

The typical CR system for FFDM is illustrated in Fig. 7.5. There are three steps. First the IP is exposure to X-rays to produce a latent image. The IP is then taken to the CR image reader or processor where a laser beam scans the IP in a systematic fashion (to render the latent image visible) and the light emitted as a result of laser scanning is collected by a light guide and sent to a digital processor which produces a CR digital image. In step three, the IP is exposed to a high-intensity light, which erases the plate thereby making it ready for another exposure.

As early as 2005, Pisano and Yaffe [7] indicated that important advantages of this system are the small size of the detector element and other advantages such as reusable image plates and varying sizes of image plates, for example. The problems of low image sharpness due to the scattering of laser light as the plate is being scanned (readout), and flat-field corrections and noise due to low light emissions, are major concerns of this technology [7].

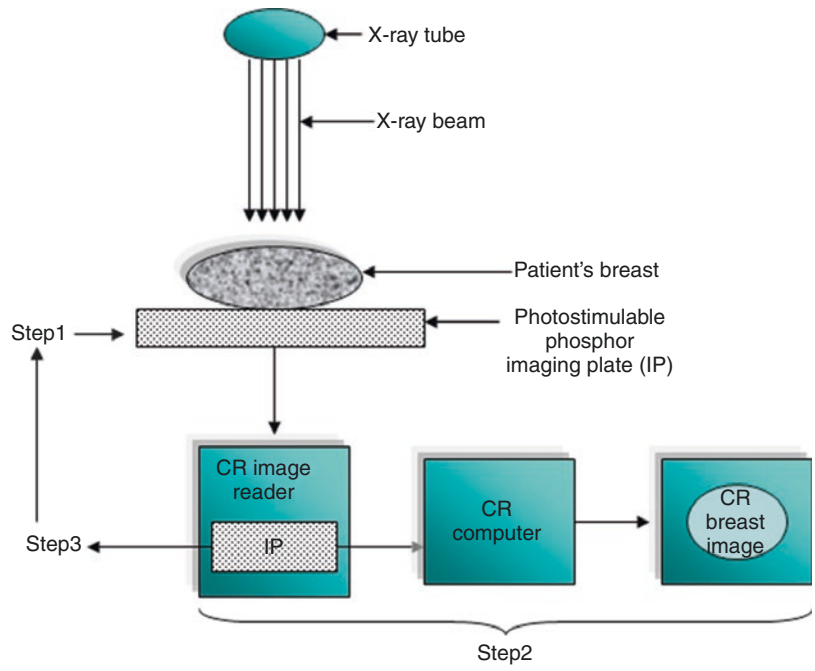

Fig. 7.5 The three fundamental steps in a typical computed radiography (CR) system for full-field digital mammography. The imaging plate (IP) is considered the digital detector in a CR imaging system

7.5.4 Flat-Panel Amorphous Selenium (a-Se) FFDM System

Yet another type of FFDM system (type 4) commercially available for clinical use is the one that is based on the *flat-panel a-Se detector* system. It is clear that the major difference between this system and the others mentioned above is that the components of this system allow for a direct conversion of X-rays to electrical signals eliminating the light conversion process. These FFDM systems are referred to as direct conversion systems.

The detector system consists of a thin layer of a photoconductor, a-Se, deposited onto a readout layer of electronics. This layer (laid down upon a glass support) is a-Si similar to the phosphor flat-panel system described above; "however, the photodiodes are replaced by a set of simple electrode pads to collect the charge" [7]. When X-rays fall upon the a-Se layer, an electric charge is released and readout using an electric field. The electrical signal generated is then digitized, and the digital data is sent to a digital computer for processing. The result of processing is a digital image of the breast.

7.6 Digital Image Post-processing Techniques

The basic concepts of digital image processing were described in detail in Chap. 2. In this section, digital image post-processing techniques specific to FFDM will be presented. Once the image is displayed on a monitor for viewing, the observer can now apply digital image *post-processing techniques* to these images for the purpose of modifying them in order to enhance diagnostic interpretation and optimize image quality in an effort to enhance diagnostic interpretation [9].

Digital image processing techniques or operations as they are often referred to can be classified into three types: point processing operations such as grayscale processing (windowing, image subtraction, and temporal averaging), local processing operations (such as spatial filtering, edge enhancement, and smoothing), and global operations such as the Fourier transform (FT). These processing algorithms can also be applied to FFDM; however there are specific processing operations meant for FFDM.

7.6.1 Specific Image Processing Algorithms for FFDM

There are several image processing algorithms developed specifically for use in FFDM systems and more importantly manufacturers have developed ones specific to their systems, and several individuals have developed others independently. Pisano, Yafee, and Kuzmiak [10], for example, have described seven of these algorithms. These include manual intensity windowing (MIW), histogram-based intensity windowing (HIW), mixture-model intensity windowing (HMIW), contrast-limited adaptive histogram equalization (CLAHE), unsharp masking, peripheral equalization, and Trex processing. While it is not within the scope of this chapter to describe the details of each of these, Pisano, Cole, Hemminger, Yaffe, Aylward, and Maidment [11] provide a comprehensive summary description of each as follows:

> *Manual intensity windowing can produce digital mammograms very similar to standard screen-film mammograms but is limited by its operator dependence. Histogram-based intensity windowing improves the conspicuity of the lesion edge, but there is a loss of detail outside the dense parts of the image. Mixture-model intensity windowing enhances the visibility of lesion borders against the fatty background, but the mixed parenchymal densities abutting the lesion may be lost. Contrast-limited adaptive histogram equalization can also provide subtle edge information but might degrade performance in the screening setting by enhancing the visibility of nuisance information. Unsharp masking enhances the sharpness of the borders of mass lesions, but this algorithm may make even an indistinct mass appear more circumscribed. Peripheral equalization displays lesion details well and preserves the peripheral information in the surrounding breast, but there may be a flattening of the image contrast in the nonperipheral portions of the image. Trex processing allows visualization of both lesion detail and breast edge information but reduces image contrast.*

7.7 Applications of FFDM

There are several applications of FFDM that have been described in the literature and which are intended to assist in not only the detection of breast cancer but also to enhance the diagnostic interpretation skills of the radiologist. These include computer-aided detection and diagnosis, digital breast tomosynthesis (DBT), and contrast-enhanced DM.

7.7.1 Computer-Aided Detection and Diagnosis

In *computer-aided detection and diagnosis* (CAD), the computer (software) is used as a tool to provide additional information to the radiologist and other related individuals in order to make a diagnosis. In other words the computer output is regarded as a "second opinion." The purpose of CAD is to improve diagnostic accuracy as well as to improve the consistency of image interpretation using the computer results as a guide.

CAD systems are essentially based on two approaches: those that use location of lesions by using the computer to search for abnormal patterns and those that quantify the image features of normal and/or abnormal patterns. CAD has been applied to screening and diagnostic mammography, chest radiography, chest CT, cardiovascular system, neuroradiology, virtual colonography, pediatric radiology, and musculoskeletal system.

There are three major technical components of a CAD system: image processing, quantitation of image features, and data processing, as shown in Fig. 7.6. It is not within the scope of this chapter to describe these in any detail; however, the following brief description of each is in order.

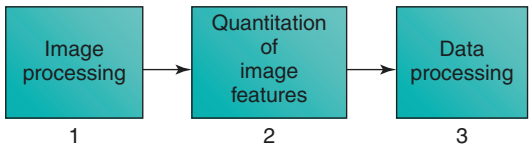

Fig. 7.6 Three major components of a CAD system for FFDM; image processing, quantitation of image features, and data processing

7.7.1.1 Image Processing

The computer uses image processing algorithms for enhancement and extraction of lesions. This process allows the computer (not the human) to pick up initial lesions and suspicious patterns. To perform this task, a number of processing operations have been used such as filtering, Fourier analysis, artificial neural networks, wavelet transform, etc. to enhance and extract lesions.

7.7.1.2 Quantitation of Image Features

Quantitation involves at least three steps to distinguish between lesions and normal anatomical structures. The first step in quantifying features in the image is to select size, contrast, and shape of candidates. The second step uses image features that have been used by radiologists for years, because diagnostic accuracy is generally high and reliable. Quantitation finds unique features that can be readily distinguished reliably between a lesion and other normal anatomical structures.

7.7.1.3 Data Processing

The third component of CAD systems is data processing. Data processing uses techniques such as rule-based methods and other approaches such as discriminant analysis, artificial neural networks, and decision tree method to distinguish between normal and abnormal patterns based on features obtained in quantitation. These methods are beyond the scope of this text and will not be described here.

7.7.2 Digital Tomosynthesis

Digital tomosynthesis (DT) uses the principles of conventional tomography to produce tomographic images that are intended to enhance the conspicuity of lesions by blurring out structures above and below the layer of interest. This technique was approved by the US Food and Drug

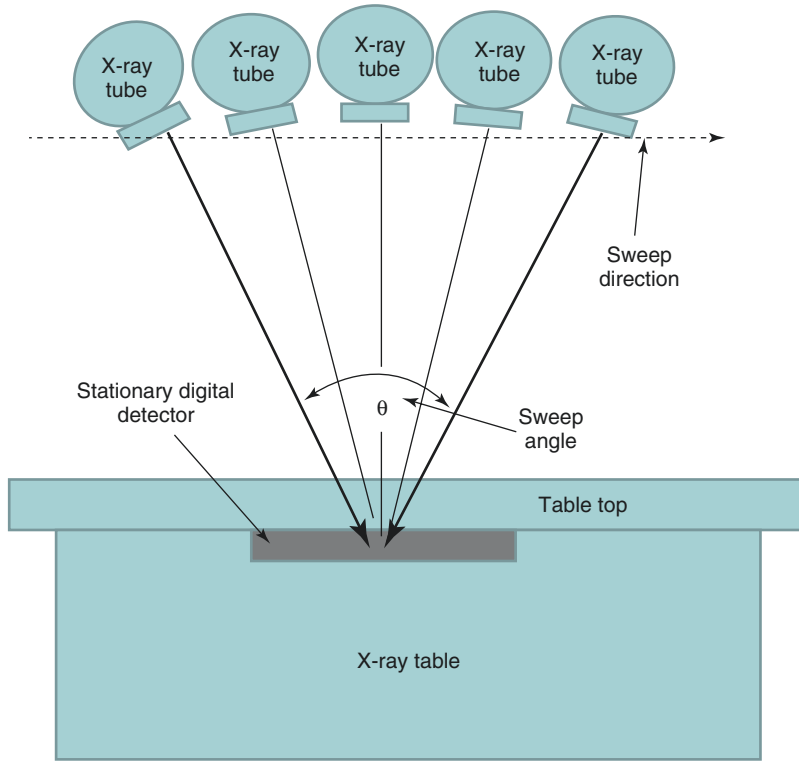

Fig. 7.7 The basic principles of digital tomosynthesis. The X-ray tube moves through various angles (θ = sweep angle), while the digital detector remains stationary. Subsequently, images are reconstructed using special image reconstruction algorithms and digital image processing to create digital tomosynthesis images and enhance visualization of lesions

Administration (USFDA) when it is used with 2D or synthesized 2D mammography.

The basic principles of digital tomosynthesis are illustrated in Fig. 7.7. Data acquisition involves movement of the X-ray tube through various angles (sweep angles), while the digital detector remains stationary. Subsequently, image reconstruction algorithms (filtered back projection or iterative reconstruction algorithms) and digital image processing are used not only create digital tomosynthesis images but enhance the visualization of image features, respectively [12]. FFDT has found clinical applications in breast {digital breast tomosynthesis (DBT)} chest, head, and neck imaging as well as in orthopaedic, emergency, and abdominal imaging [13, 14]. Basic principles and specific applications will be reviewed in more detail in Chap. 8.

7.7.2.1 Digital Breast Tomosynthesis: Three-Dimensional (3D) Mammography

Digital breast tomosynthesis (DBT) also referred to as three-dimensional (3D) mammography is a relatively new technique. DBT "improves interpretative performance and will likely replace conventional mammography in clinical practice" [15]. DBT is "a limited-angle tomographic breast imaging technique, was developed to overcome tissue superposition and its clinical adaptation was facilitated by the development of digital detectors. In DBT predefined trajectory, typically an arc spanning

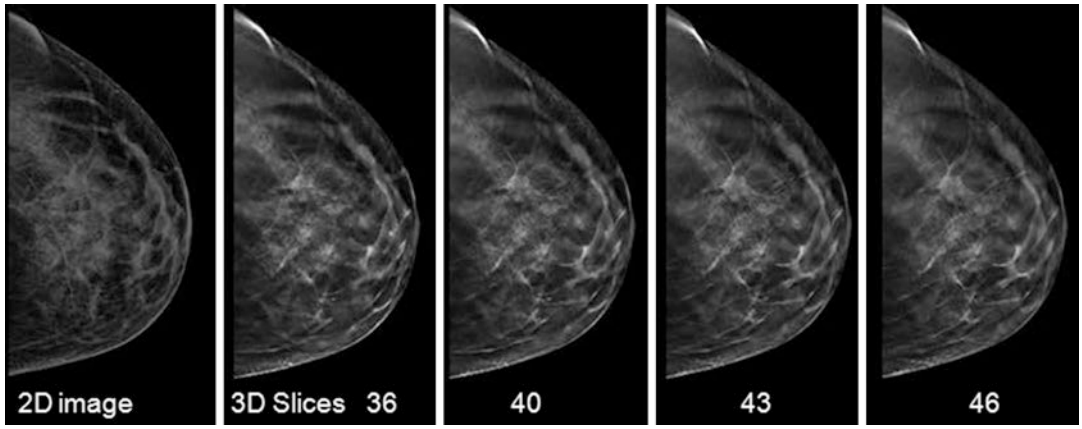

Fig. 7.8 A visual comparison of 2D and 3D mammographic images. The 3D mammographic slices reveal an area of architectural distortion, histologically proven to be cancer, which is not visible on the 2D mammogram (Legend and images-Courtesy of Hologic)

multiple projection views are acquired while the X-ray source traverses along a predefined trajectory, typically an arc spanning an angular range of 60° or less, and the acquired projection views are reconstructed to provide sections parallel to the breast support" [16]. Examples of 2D and 3D digital mammography images are shown in Fig. 7.8. DBT will be described further in Chap. 8 in terms of physical principles including reconstruction algorithms.

7.7.3 Contrast-Enhanced DM

DM can be used in angiogenesis (appearance of new vasculature in a tumour) by using iodinated contrast medium. This technique has been described by Pisano and Yaffe [10] who point out that after obtaining a mask (scout) image of the breast, several iodinated contrast images (contrast medium-enhanced) are recorded. Using computer processing, the mask image is subtracted from the contrast medium-enhanced images to produce another set of images "showing the conspicuity of the uptake of iodine in the vicinity of the lesion" [10].

There are two methods:

1. Temporal contrast-enhanced DM (CEDM).
2. Dual-energy CEDM.

"With CEDM, a DM unit has been modified to maximize the sensitivity of the unit to low concentrations of iodine contrast by increasing the voltage. The breast is in compression in a standard projection. A mask image is obtained, and following the injection of contrast, sequential or temporal images are then obtained. The examination takes approximately 15 min, and the total dose is similar to a conventional single-view mammogram. In dual-energy CEDM, the DM unit is adapted to generate high energy exposures for energies above the energy absorption of iodine (K-edge), taking advantage of the differences in X-ray attenuation through materials of different composition. Contrast is injected with the breast in compression, and two paired exposures are obtained: one set at low energy and one at high energy. This technique tends to limit problems with patient motion, as the patient is not in compression as long as with CEDM, and two views of each breast are

obtained and combined to enhance contrast-uptake areas [1]."

References

1. Bushong S. Radiologic science for technologists. 11th ed. St Louis: Elsevier-Mosby; 2017.
2. Bushberg JT, Seibert JA, Leidholdt EM Jr, Boone JM. The essential physics of medical imaging. 3rd ed. Philadelphia: Lippincott Williams and Wilkins; 2012.
3. Pisano E, Gatsonis C, Hendrick E, Yaffe M, Baum J, Acharyya S, et al. Diagnostic performance of digital versus film mammography for breast-cancer screening. N Engl J Med. 2005;353:1–11.
4. Mahesh M. Digital mammography-physics tutorial for residents. Radiographics. 2004;24:1747–60.
5. Pisano E. Current status of full-field digital mammography. Radiology. 2000;214:26–8.
6. James JJ. The current status of digital mammography. Clin Radiol. 2004;59:1–10.
7. Pisano ED, Yaffe MJ. Digital mammography. Radiology. 2005;234:353–62.
8. Samei E. Technological and psychological considerations for digital mammography displays. Radiographics. 2005;25:491–501.
9. Seeram E, Seeram D. Image postprocessing in digital radiology: a primer for technologists. J Med Imaging Radiat Sci. 2008;39(1):23–43.
10. Pisano E, Yaffe MJ, Kuzmiak CM. Digital mammography. Philadelphia, PA: Lippincott Williams & Wilkins; 2004.
11. Pisano E, Cole E, Hemminger B, Yafee M, Aylward S, Maidment A, et al. Image processing algorithms for digital mammography: a pictorial essay. Radiographics. 2000;20:1479–91.
12. Yaffe MJ, Mainprize JG. Digital Tomosynthesis. Radiol Clin North Am. 2014;52:489–97.
13. Patterson SK, Roubidoux MA. Update on new technologies in digital mammography. Int J Womens Health. 2014;6:781–8.
14. Machida H, Yuhara T, Tamura M, Ishikawa T, Tate E, Ueno E, Nye K, Sabol JM. Whole-body clinical applications of digital Tomosynthesis. Radiographics. 2016;36:735–50.
15. Hooley RJ, Durand MA, Philipotts LE. Advances in digital breast Tomosynthesis. Am J Roentgenol. 2017;208:256–66.
16. Vedantham S, Karellas A, Vijayaraghavan GR, Kopans DB. Digital breast Tomosynthesis: state of the art. Radiology. 2015;277(3):663–84.

Digital Tomosynthesis

8

Abstract

The purpose of this chapter is to present a brief description of digital tomosynthesis, a three-dimensional (3D) imaging technique that overcomes the problems of conventional two-dimensional (2D) tomography. The technique involves image acquisition, image reconstruction, and image display and communication. While image acquisition is such that the X-ray tube rotates through a limited angle about the detector which is often stationary, to obtain a number of projection data are taken from different angles. These data are subsequently reconstructed to produce individual slices of the volume of tissue scanned, using algorithms specially developed for tomosynthesis. There are two types of designs for image acquisition in digital tomosynthesis: step-and-shoot system and continuous scan system. While, in the former method, the X-ray tube that moves to every angular position stops, an exposure is taken and the tube then moves to the next angular position; in the latter system, the X-ray tube moves during the scanning of the object. Digital tomosynthesis (DT) is being applied to general radiographic imaging and to digital breast imaging referred to as digital breast tomosynthesis (DBT).

Major imaging system components include the X-ray tube and housing designed to rotate during the data acquisition, collimation and filtration, breast support, breast compression device, and either a full-field indirect flat-panel digital detector {amorphous silicon (a-Si) cesium iodide (CsI)} or a full-field direct flat-panel digital detector (a-Selenium). Furthermore DT is characterized by several parameters such as the sweep angle, sweep direction, patient barrier-object distance, number of projections, and total radiation dose. Additionally an overview of image reconstruction methods of DT, image display and communication, and radiation dose considerations is presented. Finally, this chapter concludes with an outline of synthesized 2D digital mammography (DM) and clinical applications of DT.

8.1 Introduction

One of the major limitations of projection radiography is the superimposition of all structures on the detector which makes it difficult and sometimes impossible to distinguish a particular detail. To overcome this problem, multiple views such as laterals and obliques can be taken to localize a structure; however the problem of superimposition in radiography still persists. Later, this problem can somewhat be overcome with the use of conventional tomography [1, 2].

The word *tomography* is not new. It can be traced back to the early 1920s, when a number of investigators were developing methods to image a specific layer or section of the body. At that time,

E. Seeram, *Digital Radiography*, https://doi.org/10.1007/978-981-13-3244-9_8

terms such as "body section radiography" and "stratigraphy" (from *stratum*, meaning "layer") were used to describe the technique. In 1935 Grossman refined the technique and labeled it tomography (from the Greek *tomos*, meaning "section") [2]. A conventional tomogram is an image of a section of the patient that is oriented parallel to the film. When the X-ray tube and film are moved simultaneously in opposite directions, unwanted sections can be blurred, while the desired layer or section is kept in focus. The immediate goal of tomography is to eliminate structures above and below the focused section or the focal plane. However, this is difficult to achieve and, under no circumstances, can all unwanted planes be removed. The limitations of tomography include persistent image blurring that cannot be completely removed, degradation of image contrast because of the presence of scattered radiation created by the open geometry of the X-ray beam, and other problems resulting from film-screen combinations.

Later in 1972, Grant [3] defined yet another term, "*tomosynthesis*" derived from the Greek language "tomos" meaning a slice or a section and "synthesis" meaning a process to combine or blend ideas to produce a new system. In a paper entitled "The Future of Medical Imaging," Maidment [4] identified *digital tomosynthesis* as one of his visions of the future. While conventional tomography produces one section at a time during one acquisition, digital tomosynthesis results in volume data acquisition using a single linear motion of the X-ray tube and tomographic reconstruction of the digital data from large-area flat-panel detector generating multiple images [5]. In Fig. 8.1, conventional mammography produces two-dimensional (2D) images (produced by superimposed structures) that make it difficult to clearly visualize specific pathologies of interest to the observer. This problem is solved by digital tomosynthesis (DT), a three-dimensional (3D) imaging technique that separates out the superimposed tissues, as illustrated in the right image of Fig. 8.1. Furthermore images that clearly demonstrate this advantage are shown in Fig. 8.2.

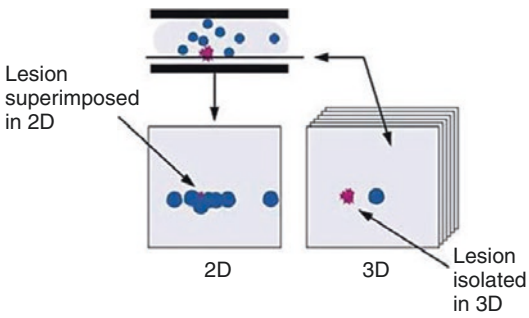

Fig. 8.1 Digital tomosynthesis produces 3D images in an effort to solve the problem of superimposition of structures which makes it difficult to visualize pathologies of interest to the observer, characteristic of conventional mammography, a 2D imaging technique (From Smith A. Design Considerations in Optimizing a Breast Tomosynthesis System. White Paper, Bedford, MA. Hologic, Inc.™ 2011. Reproduced by permission)

8.2 Digital Tomosynthesis: Definition and Principles

A more complete definition is that digital tomosynthesis "is a three-dimensional imaging technique based on the reconstruction of several planar radiographs. During the image acquisition in tomosynthesis, the X-ray tube moves around the detector which is often stationary, and a number of projection images are taken from different angles. Individual slices from the reconstructed volume can be studied. With the effective reduction of the visibility of the overlapping normal tissue, the detection of pathological lesions is improved when compared with projection radiography" [6]. Digital tomosynthesis techniques have been applied in several clinical areas including orthopedics, angiography, chest, and breast imaging. In radiographic applications (chest, skeletal, head and neck, emergency, and abdominal radiography), the technique can be referred to as radiographic tomosynthesis. In this chapter, however, the term digital tomosynthesis will be used to imply radiographic tomosynthesis. In particular, digital breast tomosynthesis (DBT) has received increasing attention in the literature and will be discussed later in this chapter. DBT is "an extension of digital mammography that produces quasi three-dimensional reconstructed images from a set of low-dose X-ray projections acquired over a limited angular range" [7].

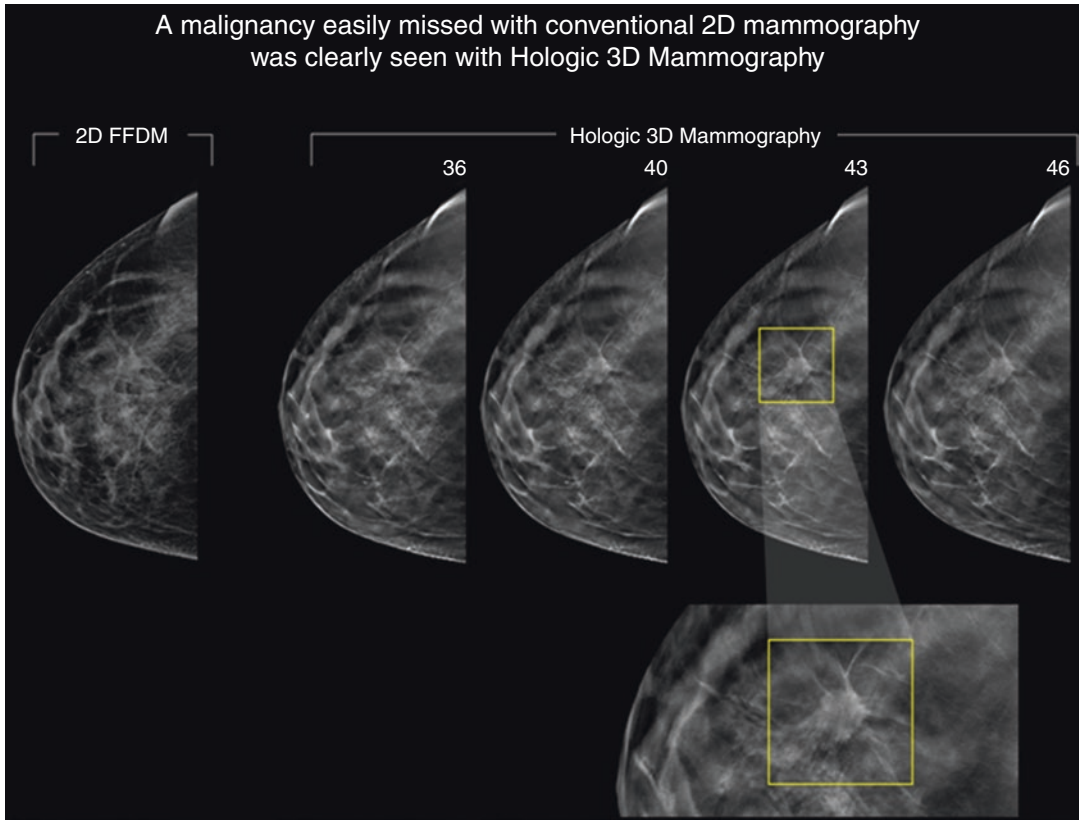

Fig. 8.2 One of the significant advantages of digital tomosynthesis is that superimposed tissues from different sections of the breast are clearly demonstrated compared to full-field digital mammography (FFDM) (Courtesy of Hologic®)

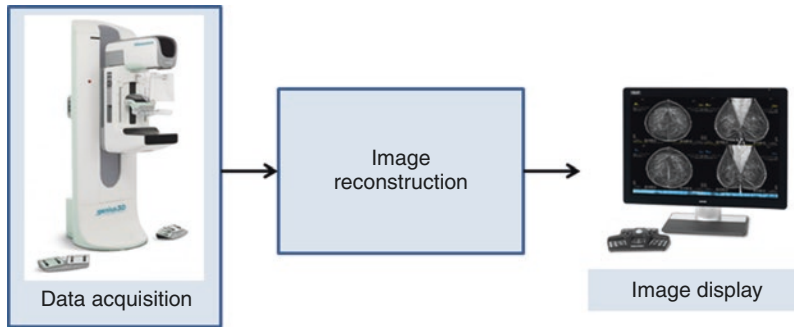

Fig. 8.3 The fundamental process of digital tomosynthesis includes at least three steps, namely, data acquisition, image reconstruction, and image display on a workstation (Images courtesy of Hologic®)

The fundamental process of digital tomosynthesis includes at least three steps, namely, image acquisition, image reconstruction, and image display on a workstation, as illustrated in Fig. 8.3. Each of these will now be briefly outlined.

8.2.1 Image Acquisition and System Components

The fundamental principles of digital tomosynthesis image acquisition are shown in Fig. 8.4.

As illustrated, the X-ray tube moves through various positions (five in this case) which describes an angle θ, the sweep angle, while the digital detector remains stationary. In this manner several slices are acquired at low doses and stored. An extension of Fig. 8.4 is shown in Fig. 8.5 which illustrates both acquisition images of the objects represented as a white square and a black circle and the resulting tomosynthesis reconstruction images in two planes.

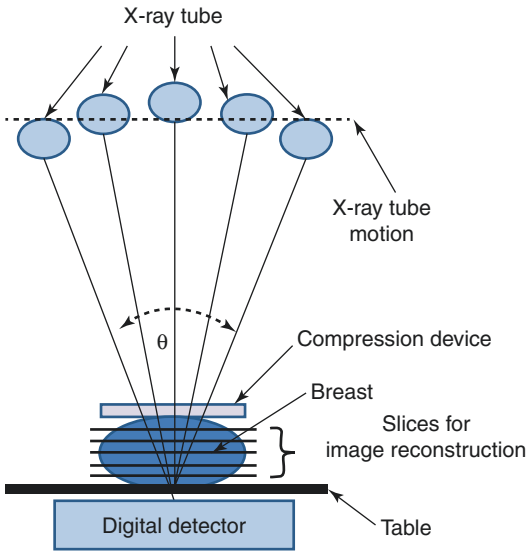

Fig. 8.4 The basic principles of digital tomosynthesis image acquisition. See text for further explanation

In Fig. 8.6, when the X-ray tube is perpendicular to the detector, structures (a lesion and ellipse) located at different heights in the breast are superimposed on that image. These structures are separated out in the non-zero degree (off axis) positions of the X-ray tube [8]. Subsequent image reconstruction and image processing by the computer render structures located at different planes in the patient clearly visible on the tomographic images as shown in Fig. 8.7. This figure shows a significant advantage of digital tomosynthesis; superimposed tissues from different sections of the breast are clearly demonstrated in the images compared to the two-dimensional full-field digital mammography (2DFFDM) or simply a two-dimensional digital mammography (2DDM) image.

There are two types of designs for image acquisition in digital tomosynthesis: (1) step-and-shoot system and (2) continuous scan system [7, 9].

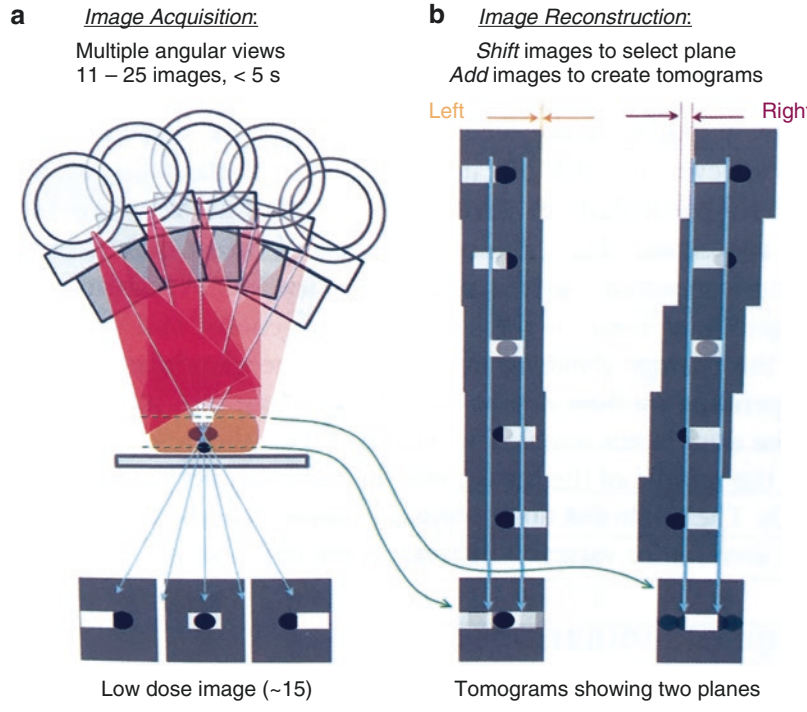

Fig. 8.5 An extension of Fig. 8.4, showing (**a**) image acquisition and (**b**) image reconstruction steps involved in DT. (From Bushberg JT, Seibert JA, Leidholdt Jr. EM and Boone JM. The Essential Physics of Medical Imaging. Philadelphia, PA, Wolters Kluwer/ Lippincott Williams & Wilkins. 2012. Reproduced by permission)

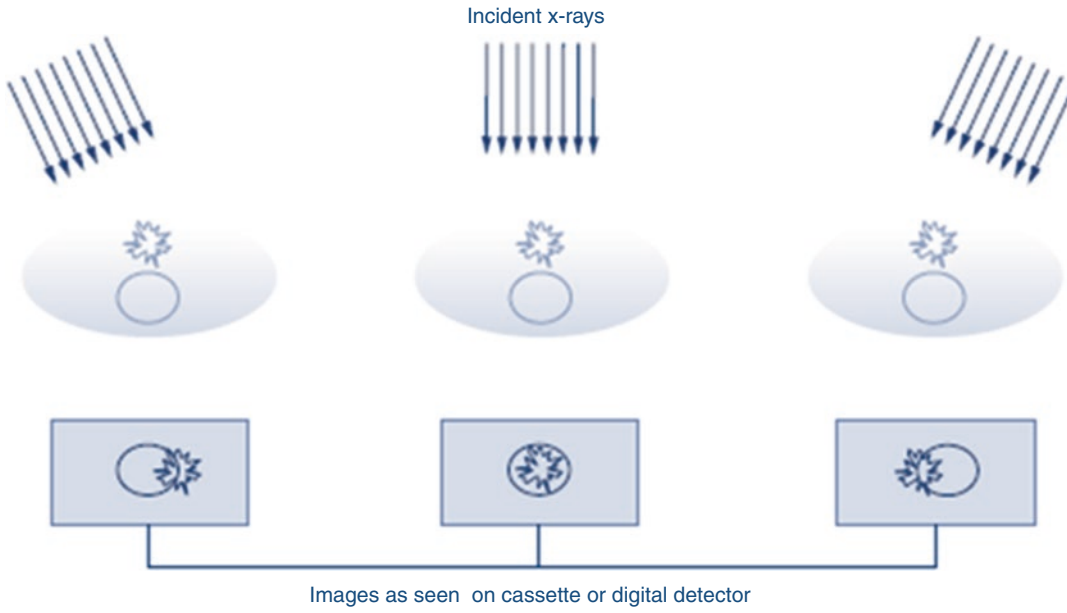

Fig. 8.6 In digital tomosynthesis imaging, images acquired from different angles separate structures at differing heights. Conventional mammography would acquire only the central image (From Smith A. Fundamentals of Breast Tomosynthesis. Improving the Performance of Mammography. White Paper, Bedford, MA. Hologic, Inc.™ 2012. Reproduced by permission)

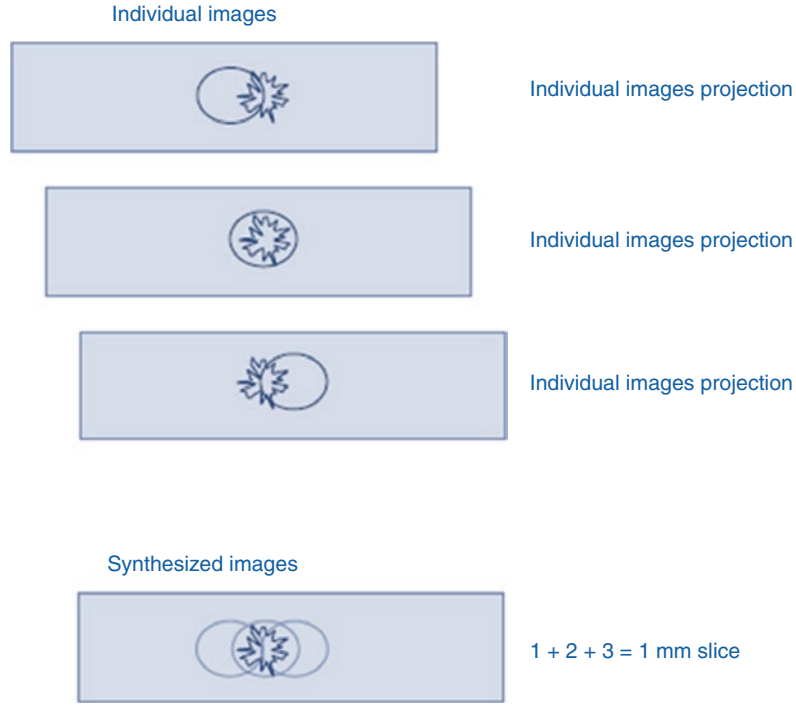

Fig. 8.7 Tomosynthesis enhances objects at a given height. By shifting and adding the acquired projections, 3D imaging increases the visibility of objects by blurring out objects from other heights (From Smith A. Fundamentals of Breast Tomosynthesis. Improving the Performance of Mammography. White Paper, Bedford, MA. Hologic, Inc.™ 2012. Reproduced by permission)

As described by Smith [9], "in step-and-shoot system, the X-ray tube moves to each angular position, then stops and takes an exposure, then moves to the next location. In a continuous scan system, the X-ray tube moves in a generally uniform velocity throughout the scan." While the former design may introduce some degree of image blurring due to the mechanics of stopping the X-ray tube (e.g., shaking of the tube), the latter design uses additional technology such as short X-ray pulse widths and thinner tube filters (compared to 2D systems) to reduce image blur [9]. Finally, Smith [9] reports that if "all factors are taken into consideration the continuous scan motion is probably preferable because of the fast scan capability, and because, despite the continuous motion, the level of pulse width blurring is clinically insignificant."

The major imaging system components of most digital tomosynthesis systems include the X-ray tube and housing designed to rotate during the acquisition of images, collimation and filtration, breast support, breast compression device, and either a full-field indirect flat-panel digital detector [amorphous silicon (a-Si) cesium iodide (CsI)] or a full-field direct flat-panel digital detector [amorphous selenium (a-Se)] as described in Chap. 7. A notable distinction between a mammography system and a tomosynthesis system as pointed out by Sechopoulos [10] is that the X-ray tube must be capable of rotation and that the readout of the data on the detector must be fast.

While one of the problems with X-ray tube motion is that of focal spot motion blurring which leads to low spatial resolution, the other is the long scanning times of the FFDM systems. To overcome these problems, a design which is based on a stationary system to replace the rotation of the X-ray tube, one that makes use of a carbon nanotube (CNT) array of X-ray sources (32 X-ray sources which span 370 mm and provide a coverage over 30°), is being investigated [10]. This technology will not be described in this chapter.

8.2.2 Image Acquisition Parameters

As in conventional tomography, which is characterized by several parameters such as object plane, tomographic angle, fulcrum, and slice thickness [11], digital tomosynthesis is also characterized by several parameters. These include the sweep angle, sweep direction, patient barrier-object distance, number of projections, and total radiation dose. The definitions of each of these as provided below by Machida et al. [5] are as follows:

1. *Sweep angle*: "The term sweep angle refers to the total arc about the center of the detector as defined by the focal spot position from the first to the final projection in the tomosynthesis acquisition or sweep. A sweep angle of 40° signifies a –20° to +20° sweep. The sweep angle can be varied between 20° and 40° on our flat-panel detector system."
2. *Sweep direction*: "The direction of X-ray tube movement relative to the object or body part of interest during a sweep. Sweep direction can be arbitrarily determined by altering the position or direction of the object or body part."
3. *Patient barrier-object distance*: "Is the minimum distance between the surface of the patient barrier and the object of interest. Detector-object distance should be used for more accurate geometric analysis, but patient barrier-object distance is convenient because it can be determined more easily and accurately."
4. *Number of projections and projection density*: "The number of projections is simply the number of X-ray projection images acquired during a single sweep. The projections are acquired at approximately constant angle intervals. For example, with a 40° sweep with 40 projections, a projection is acquired every 1°. As mentioned earlier, any number of projections from 25 to 60 can be selected on our flat-panel detector system. Projection density (number of projections divided by sweep angle) is used as an important conceptual parameter in the minimization of ripple artifact." Sechopoulos [10] reports that the research indicates that 15–20 projections proved to be optimal.
5. *Total radiation dose*: "Is the cumulative sum of the doses for all projections." Digital breast

tomosynthesis (DBT), a limited angle computed tomography technique, has shown significant promises in addressing these limitations [9, 10]. In DBT, multiple projection images are acquired at different viewing angles and reconstructed into a 3D dataset, which can be viewed in thin slices with high in-plane resolution without suffering from tissue overlap. It has the potential to improve the effectiveness of early breast cancer screening at a similar dose and cost to full-field digital mammography (FFDM) [11]. The first commercial DBT scanner received FDA approval in early 2011. Several other DBT systems from different vendors are currently under clinical trials [12–16].

8.2.3 Image Reconstruction

After image acquisition, the next major step in the DT process is image reconstruction, a process based on the use of an algorithm referred to as a reconstruction algorithm. After projection data are obtained from the object (patient), the algorithm uses the attenuation coefficients (μs) data (from the object/patient) measured by the detector to systematically build up the image of the spatial distribution of μs, for viewing and interpretation. Figure 8.8 shows the difference between projection images of the breast acquired at three angles and the reconstructed slices at three different heights for each of the three structures.

The reconstruction algorithms used in DT and DBT fall into two categories, namely, iterative reconstruction (IR) algorithms and the filtered back projection (FBP) algorithm. The workings of these algorithms are beyond the scope of this text; however, the following points are noteworthy:

1. A major problem with the FBP algorithm is that of noise in the image and streak artifacts; however sharpness and contrast can be improved with the use of digital filters.
2. IR algorithms generally make use of FBP to create simulated data which are then compared with the actual measured projection data to determine differences in image noise. Once this difference is determined, it is applied to the simulated projection to correct for inconsistencies. The system reconstructs a new image, and the process repeats until the difference between the measured and simulated projections is minor enough to be acceptable.

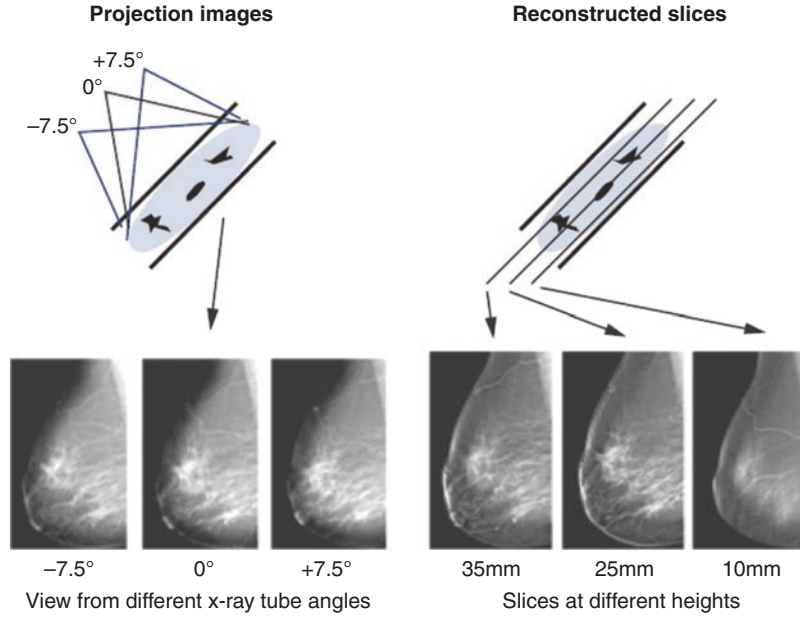

Fig. 8.8 The difference between projection images of the breast acquired at three angles and the reconstructed slices at three different heights for each of the three structures (From Smith A. Fundamentals of Breast Tomosynthesis. Improving the Performance of Mammography. White Paper, Bedford, MA. Hologic, Inc.™ 2012. Reproduced by permission)

The algorithms used in DT and DBT systems are of several types including maximum likelihood (ML belongs to a class of iterative techniques), algebraic reconstruction techniques (ART), and the filtered back projection (FBP) [9] and iterative reconstruction (IR) algorithms. These algorithms have their roots in the reconstruction algorithms used in computed tomography (CT); however, because of the "limited angle-limited view nature" characteristic of DT and DBT, that is, a smaller number of projections are obtained from the limited scanning angle (compared to CT projection data), the algorithms had to be changed to account for these differences between CT and DT and DBT [7]. Two notable algorithms in DT and DBT are the FBP and IR algorithms such as the (ART) [5, 7, 9, 12, 13] and, the more recent of IR algorithms, the DBT-ASiR (DBT-dedicated adaptive statistical iterative reconstruction) algorithm [14]. These algorithms are beyond the scope of this text, and therefore they will not be described here. For a thorough description of DBT image reconstruction algorithms, the reader should refer to the papers by Dobbins and Godfrey [15] and by Sechopoulos [16].

DBT is a kind of computed tomography (CT) where a much less number of projections acquired from a limited scanning angle are used for image reconstruction.

A summary of the main characteristics of common DT systems from DBT vendors such as Siemens Healthineers, Hologic, and General Electric (GE) Healthcare and DT vendors such as Shimadzu (for radiography and fluoroscopy) and GE Healthcare (for radiography) is provided by Machida et al. [14]. The characteristics include sweep angle, number of projections, detector technology, pixel pitch, scan time, and reconstruction algorithm. For example, the algorithm used by all vendors listed is the FBP algorithm, except for DBT (GE Healthcare) which uses an IR algorithm and Shimadzu which uses both FBP and IR algorithms for DT radiography and fluoroscopy. While the detector technology used by Siemens and Hologic for their DBT systems is the a-Se flat-panel system, it is CsI/a-Si for both the GE Healthcare DBT and DT (radiography) systems. Finally, the sweep angles vary from vendor to vendor. For example, while the sweep angle for the Siemens DBT system is 50°, it is 15° and 25° for the Hologic and GE Healthcare DBT systems, respectively [14].

Another algorithm for DBT is one described by Rodriguez-Ruiz et al. [17]. This algorithm is called EMPIRE (enhanced multiple parameter iterative reconstruction). This is fundamentally FBP with iterative optimizations. The performance of EMPIRE was compared with the performance of the FBP algorithm using human observers and machine learning algorithms. The results of this study showed that "the new algorithm provides DBT volumes with better contrast and image quality, fewer artifacts, and improved visibility of calcifications for human observers, as well as improved detection performance with deep-learning algorithm" [17].

8.2.4 Image Display and Communication

Once DT and DBT images are acquired and processed, they are subsequently displayed on a monitor designed for DT and DBT image display. Display monitors for digital mammography should be approved by the Food and Drug Administration (FDA) and must be calibrated, and have at least five million pixels, and meet luminance standards of the American College of Radiology (ACR). A typical dedicated DT workstation and monitors are shown in Fig. 8.9. Furthermore, in order to provide fast access to DT images, a picture archiving and communication system (PACS) with digital imaging and communications in medicine (DICOM) standard is also an integral component of DT systems [9, 14]. Furthermore, the reconstructed images can be displayed and viewed in sequence using a

Fig. 8.9 A dedicated workstation including monitors used for DT systems (Courtesy of Hologic®)

cine-loop technique, or they can be displayed and viewed as single images.

The interested reader should explore the different DT vendor websites for details of the performance characteristics of their workstations. Commercial DBT systems are available from vendors such as Hologic®, General Electric Healthcare, Siemens Healthineers, Fijufilm, and Giotto.

8.3 Radiation Dose Considerations

As mentioned above, Machida et al. [5] noted that one of the parameters of importance in digital tomosynthesis is the total radiation dose. Dose in breast imaging is critical since the glandular tissue is the "site of carcinogenesis" [18]. Therefore the mean glandular dose (MGD) is the metric used in breast imaging dose studies, and it has been estimated by Monte Carlo simulation [10, 15]. Dosimetry for mammography has been established and described by Bushberg et al. [18]; however, as described earlier, in DBT, the X-ray tube is moving with either a stationary detector or a moving detector, and this approach to image acquisition poses more of a challenge in developing DBT dosimetry. For this reason DBT dosimetry methodology will not be described in this chapter. For details of radiation dosimetry in breast imaging, including dosimetry protocols, the interested reader should refer to Bushberg et al. [18], Sechopoulos [10], and Vedantham et al. [13]

Vedantham et al. [13] reported two studies examining the dose from FFDM and DBT systems. The first study showed that doses from a FFDM system and a DBT system were measured and compared, and the results showed that the MGD was higher for the DBT system compared to the FFDM system. In addition, for a combined DBT-FFDM mode, the results showed that the MGD was lower than the regulatory limit of 3 mGy for "a CC-equivalent view of an average breast" using the US dosimetry protocol. Furthermore, in a dose review estimates in DBT relative to two-view FFDM, examples of the results of several dose studies have also been reported by Vedantham et al. [13] Finally Table 8.1 summarizes the conclusions of five dose studies in DBT ranging from 2012 to 2018.

Table 8.1 The quoted findings of five studies examining the dose from DBT

Year	Authors	Title of study	Quoted findings
2018	Gennaro et al. [19]	Radiation dose with digital breast tomosynthesis compared to digital mammography: per-view analysis	Our findings show *a modest increase of radiation dose to the breast by tomosynthesis* compared to FFDM. Given the emerging role of DBT, its use in conjunction with synthetic 2D images should not be deterred by concerns regarding radiation burden and should draw on evidence of potential clinical benefit
2017	Alakhras et al. [20]	Relationship between radiation dose and image quality in digital breast tomosynthesis	*DBT dose was 13% higher than DM*, but this differential is highly dependent on thickness. Visibility of masses was equal to a reference image (produced at 100% dose) when dose was reduced by 75 and 50% for DBT and DM. For microcalcifications, visibility was comparable with the reference image for both modalities at 50% dose. This study highlighted the potential for reducing dose with DBT
2017	James et al. [21]	Breast Radiation Dose With Contrast Enhanced Spectral Mammography (CESM) Compared With 2D FFDM and 3D Tomosynthesis Mammography	*CESM was found to increase AGD* at a mean breast thickness of 63 mm by approximately 0.9 mGy and 0.5 mGy compared with 2D FFDM and 3D tomosynthesis, respectively. Of note, CESM provides a standard image (similar to 2D FFDM) that is obtained using the low-energy projection. *Overall, the AGD from CESM falls below the dose limit of 3 mGy set by Mammography Quality Standards Act regulations*
2016	Gilbert et al. [22]	Digital breast tomosynthesis (DBT): a review of the evidence for use as a screening tool	*Adding DBT to FFDM more than doubles the radiation dose, a woman would receive in routine breast screening.* As mentioned previously, there is an opportunity to use synthetic 2D images in combination with DBT instead of conventional FFDM with only a slight increase in dose compared with FFDM and DBT. *Once synthetic 2D images have been shown as an acceptable alternative, the marginal increase in radiation dose becomes much less of an issue*
2015	Svahn et al. [23]	Review of radiation dose estimates in digital breast tomosynthesis relative to those in two-view full-field digital mammography	*Stand-alone DBT operated at lower to slightly higher radiation doses in comparison with FFDM*. For DBT combined with FFDM, radiation doses were elevated, at maximum by a factor ~2 1/4 of that of FFDM alone. In this setting, a replacement of FFDM with synthetic 2D-views reduced the breast dose approximately by half, which has substantial implications for population screening programs
2012	Olgar et al. [24]	Average glandular dose in digital mammography and breast tomosynthesis	*The mean calculated AGD per exposure in 3D imaging mode was on average 34% higher than for 2D imaging* mode for patients examined with the same mean compressed breast thickness (CBT)

8.4 Synthesized 2D Digital Mammography

Two-dimensional full-field digital mammography (2DFFDM) system components and imaging technology were described in detail in Chap. 7. The introduction and subsequent development and implementation of DBT were approved by the FDA in 2011 with the restriction that "all DBT acquisitions must be paired with a 2D image to assure adequate interpretative information is provided" [25]. Such pairing of a 2D FFDM image with a DBT image is illustrated in Fig. 8.10 and noted as a "Combo" mode. An important technical advance in DBT is the production of a synthetic 2D image which can be generated from the DBT data set. This operation is referred to as synthesized 2D digital mammography. Additionally, the pairing of the DBT image with the synthesized 2D image (C-View image) is noted as a

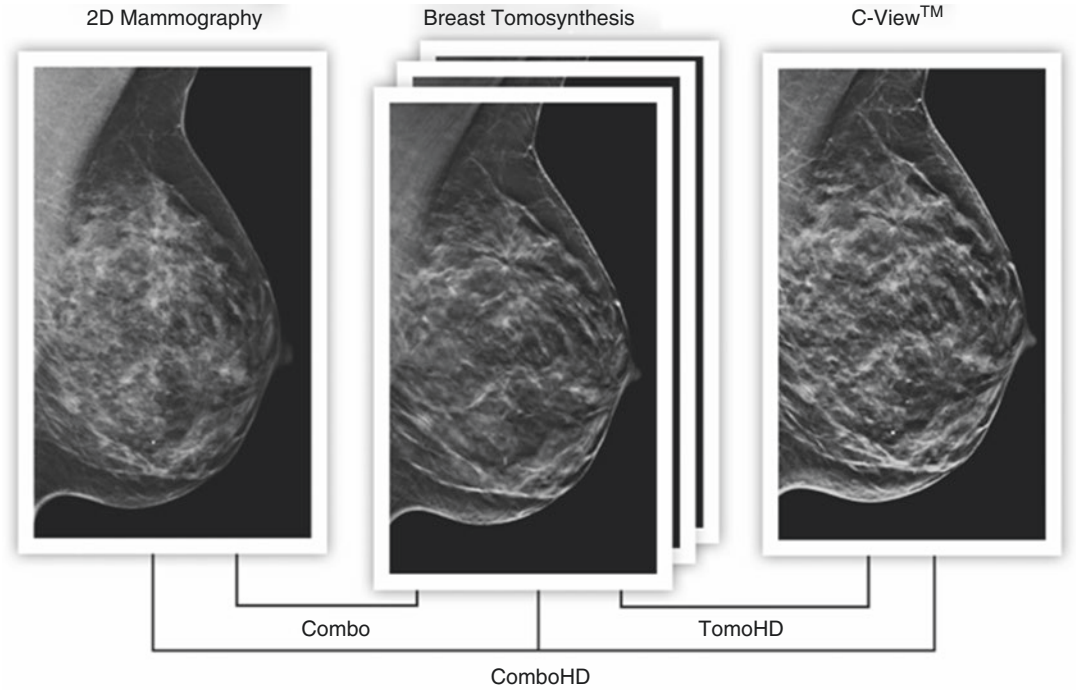

Fig. 8.10 Three modes of combining images in DBT (See text for further explanation. Courtesy of Hologic®)

"TomoHD" image (Fig. 8.10). If all three of the images are combined, this is referred to as "ComboHD" mode (Fig. 8.10). In a study entitled "Two-View Digital Breast Tomosynthesis Screening with Synthetically Reconstructed Projection Images: Comparison with Digital Breast Tomosynthesis with Full-Field Digital Mammographic Images" [26], the researchers concluded that the combination of the synthesized 2D images (C-View image) with the DBT image (TomoHD mode) proved to be better that the 2D mammography alone. As a result of several studies on the efficacy of synthesized 2D images in DBT, Zuckerman et al. [27] indicate that synthesized 2D digital mammography is gaining acceptance as a new tool to image the breast coupled with DBT in an effort to "reduce radiation dose and maintain screening outcomes."

This has created a need for synthesized 2D images, and the result is that breast imaging vendors are now engaged in providing the software needed to generate these 2D synthesized images. Hologic®, GE Healthcare, and Siemens Healthineers are examples of these vendors. Hologic® has developed algorithms known as C-View™ which has been approved by the FDA after clinical validation, to be useful in replacing 2DFFDM images in screening [28]. While GE Healthcare is developing their product referred to as V-View, Siemens is also working on their product referred to as Insight 2D, at the time of this writing. Therefore only the C-View™ will be highlighted here to provide an overview of this technology.

The process for generating a synthesized 2D image from the DBT image data is broadly illustrated in Fig. 8.11. First, DBT data acquisition is performed followed by the reconstruction of DBT 3D images. Next the DBT 3D images are stacked and subsequently subjected to a synthesized 2D algorithm (C-View™ algorithm for example) to generate synthesized 2D images. For additional details of the goal/design objectives and description of the C-View algorithm, the interested reader should refer to the work of Smith [28]. Figure 8.12 shows the visibility of

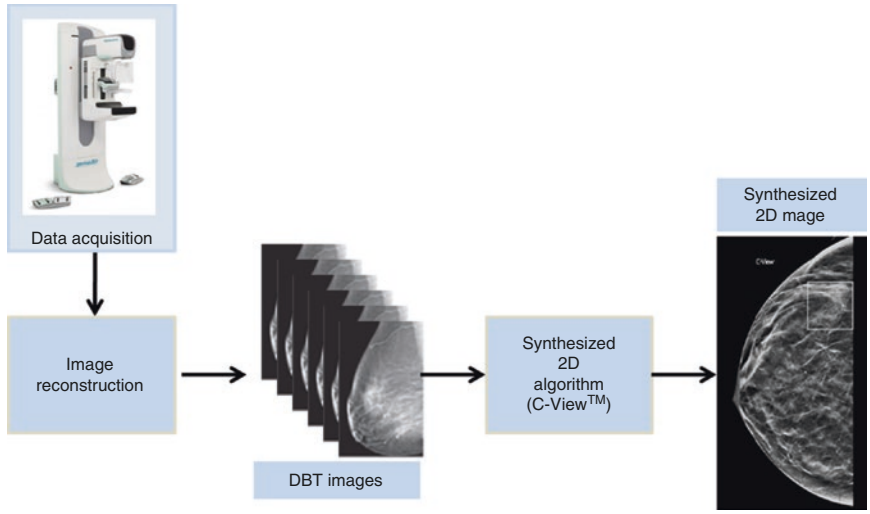

Fig. 8.11 The fundamental steps for generating a synthesized 2D image from the DBT image data (See text for further explanation. Images Courtesy of Hologic®)

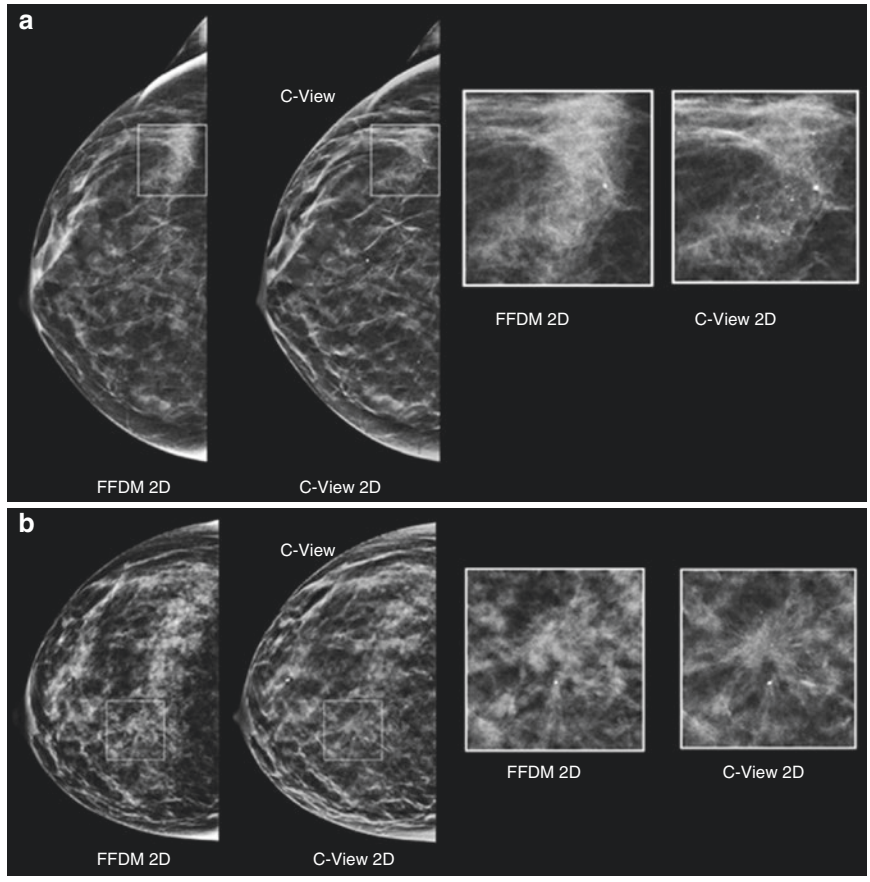

Fig. 8.12 The visibility of calcifications (**a**) and invasive ductal carcinoma (**b**) on 2D FFDM compared with that of the synthesized 2D image (From Smith A. Synthesized 2D mammographic Imaging. White Paper, Bedford, MA. Hologic, Inc.™ 2016. Reproduced by permission)

calcifications on 2D FFDM compared with that of the synthesized 2D image. Furthermore, clinical studies validating the safety and efficacy of synthesized 2D (C-View™) particularly as a replacement of 2D FFDM imaging are also summarized by Smith [28] who lists studies from 2014 to 2015 published in *Radiology, European Radiology* and those presented at the several annual general conferences of the Radiological Society of North America (RSNA).

8.5 Clinical Applications

In an article by Machida et al. [14] published in *RadioGraphics,* the authors present a comprehensive review of the clinical applications of DT. They specifically outline applications in DBT, applications of DT in head and neck imaging, chest and abdominal imaging, orthopedic imaging, and emergency imaging. Furthermore they also present an overview of clinical utilities and limitations. This review showed that [14]:

1. DT is useful in imaging the breast, chest, head and neck, skeletal system emergency, and abdomen.
2. When comparing DT with conventional mammography, radiography, and fluoroscopy, "DT can improve detection of breast cancer, pulmonary nodules, sinonasal mucosal thickening, and bone fractures and delineation of complex anatomic structures such as the ostiomeatal unit, atlantoaxial joint, carpal and tarsal bones, and pancreatobiliary and gastrointestinal tracts" [14].
3. In a comparison with computed tomography, "DT offers reduced radiation exposure, better in-plane resolution to improve assessment of fine bony changes, and less metallic artifact, improving postoperative evaluation of patients with metallic prostheses and osteosynthesis materials" [14].
4. In terms of patient positioning, "DT is also useful for functional, weight-bearing, and stress tests. To optimize patient management, a comprehensive understanding of the clinical applications and limitations of whole-body DT applications is important for improvement of diagnostic quality, workflow, and cost-effectiveness" [14].

References

1. Vallebona A. Radiography with great enlargement (microradiography) and a technical method for radiographic dissociation of the shadow. Radiology. 1931;17:340–1.
2. Bocage E. M. Patent No. 536, 464, Paris. History of tomography. Medicamundi, 1974; 19: 106–115.
3. Grant DG. Tomosynthesis: a three-dimensional radiographic imaging technique. IEEE Trans Biomed EngBME. 1972;19:20–8.
4. Maidment ADA. The future of medical imaging. Radiat Prot Dosim. 2010;139(1–3):3–7.
5. Machida H, Yuhara T, Mori T, Ueno E, Moribe Y, Sabol JM. Whole-body clinical applications of digital Tomosynthesis. Radiographics. 2010;30:549–62.
6. Tingberg A. X-ray Tomosynthesis: a review of its use for breast and chest imaging. Radiol Clin N Am. 2010;52:489–97.
7. Yaffe MJ, Mainprize JG. Digital Tomosynthesis. Technique Radiol Clin North Am. 2014;52:489–97.
8. Smith A. Fundamentals of breast tomosynthesis. Improving the performance of mammography. White Paper, Bedford, MA. Hologic, Inc™. 2012.
9. Smith A. Design considerations in optimizing a breast tomosynthesis system. White Paper, Bedford, MA. Hologic, Inc™. 2011.
10. Sechopoulos I. A review of breast tomosynthesis. Part I. the image acquisition process. Med Phys. 2013;40(1):014301-1–014301-12.
11. Bushong S. Radiologic science for technologists. 11th ed. St Louis: MO. Elsevier; 2017.
12. Gomi T. A comparison of reconstruction algorithms regarding exposure dose reductions during digital breast Tomosynthesis. J Biomed Sci Eng. 2014;7:516–25.
13. Vedantham S, Karellas A, Vijayaraghavan GR, Kopans DB. Digital breast Tomosynthesis: state-of-the-art. Radiology. 2015;277:663–84.
14. Machida H, Yuhara T, Tamura M, Ishikawa T, Tate E, Ueno E, Nye K, Sabol JM. Whole-body clinical applications of digital Tomosynthesis. Radiographics. 2016;36:735–50.
15. Dobbins JT 3rd, Godfrey DJ. Digital X-ray tomosynthesis: current state of the art and clinical potential. Phys Med Biol. 2003;48(19):R65–R106.
16. Sechopoulos I. A review of breast tomosynthesis. Part II. Image reconstruction, processing and analysis, and advanced applications. Med Phys. 2013;40(1):014302.
17. Rodriguez-Ruiz A, Teuwen J, Vreemann S, Bouwman RW, van Engen RE, Karssemeijer N, Mann RM, Gubern-Merida A, Sechopoulos I. New reconstruc-

tion algorithm for digital breast tomosynthesis: better image quality for humans and computers. Acta Radiol. 2017;0(0):1–9.
18. Bushberg JT, Seibert JA, Leidholdt EM Jr, Boone JM. The essential physics of medical imaging. Philadelphia, PA: Wolters Kluwer/Lippincott Williams & Wilkins; 2012.
19. Gennaro G, Bernardi D, Houssami N. Radiation dose with digital breast tomosynthesis compared to digital mammography: per-view analysis. Eur Radiol. 2018;28(2):573–81.
20. Alakhras MM, Mello-Thoms C, Bourne R, Rickard M, Diffey J, Brennan PC. Relationship between radiation dose and image quality in digital breast tomosynthesis. Radiat Prot Dosim. 2017;173(I4):351–60.
21. James JR, Pavlicek W, Hanson JA, Boltz TF, Patel BK. Breast radiation dose with CESM compared with 2D FFDM and 3D Tomosynthesis mammography. AJR Am J Roentgenol. 2017;208:2–9.
22. Gilbert FJ, Tucker L, Young KC. Digital breast tomosynthesis (DBT): a review of the evidence for use as a screening tool. Clin Radiol. 2016;71:141–50.
23. Svahn TM, Houssami N, Sechopoulos I, Mattsson S. Review of radiation dose estimates in digital breast tomosynthesis relative to those in two-view full-field digital mammography. Breast. 2015;24(2):93–9.
24. Olgar T, Kahn T, Gosch D. Average glandular dose in digital mammography and breast tomosynthesis. Rofo. 2012;184(10):911–8.
25. Nelson JS, Wells JR, Baker JA, Samei E. How does c-view image quality compare with conventional 2D FFDM? Med Phys. 2016;43(5):2538–47.
26. Skaane P, Bandos A, Eben E. Two-view digital breast Tomosynthesis screening with synthetically reconstructed projection images: comparison with digital breast Tomosynthesis with full-field digital mammographic images. Radiology. 2014;271(3):655–63.
27. Zuckerman SP, Maidment ADA, Weinstein SP, McDonald ES, Conant EF. Imaging with synthesized 2D mammography: differences, advantages, and pitfalls compared with digital mammography. AJR Am J Roentgenol. 2017;209(1):222–9.
28. Smith A. Synthesized 2D mammographic imaging. White Paper, Bedford, MA. Hologic, Inc™. 2016.

Picture Archiving and Communication Systems

9

Abstract

This chapter described the major system components and technologies related to picture archiving and communication systems (PACS) used in medical imaging. First a definition of PACS is provided followed by a brief description of the major components of a PACS. Several definitions have been cited in the chapter. One such definition is that PACS is fundamentally a highly sophisticated storage system that stores data and images from the image acquisition equipment and uses computer networks and application software not only to display images for viewing and interpretation but to transmit the data and images to remote locations, and as a result, PACS has become an enterprise image management system. Major components include image acquisition modalities, computer networks, the PACS main computer, image storage characteristics, image compression, display and analysis workstations, the radiology information system (RIS) and the PACS broker, and finally the web server. PACS is also characterized by the use of two communication standards, Digital Imaging and Communications in Medicine (DICOM®) and Health Level-7 (HL-7). While HL-7 deals with the flow of textual information between the hospital information system (HIS) and the RIS and other information systems in the hospital, DICOM® deals primarily with the exchange of images in the radiology department and facilitates communication among manufacturer-specific imaging equipment.

DICOM® is an international standard to transmit, store, retrieve, print, process, and display medical imaging information. An essential feature of PACS is that in order to facilitate communications between various computer-based healthcare information systems and imaging modalities and PACS vendors through the integrated use of DICOM® and HL-7 requires an initiative referred to as Integrating the Healthcare Enterprise (IHE). IHE provides a *process flowchart* that illustrates the resources and tools such as standards, technical specifications, integration profiles, and technical frameworks, for example, for the purpose of helping users not only to integrate systems but also to share information in an efficient and effective manner.

Two other significant features gaining widespread attention in the PACS environment is the vendor neutral archive (VNA) and more recently enterprise imaging. A VNA is an archive that "can be easily migrated, ported to interface with another vendor's viewing, acquisition, and workflow engine to manage medical images and related information"; enterprise imaging is a technology that offers improved efficiencies in data archiving and management and creates a comprehensive program that can combine data from different departments into one system. An enterprise imaging system should consist of elements such as governance,

E. Seeram, *Digital Radiography*, https://doi.org/10.1007/978-981-13-3244-9_9

enterprise imaging strategy, enterprise imaging platform (infrastructure), clinical images and multimedia content, electronic health record (EHR) enterprise viewer, image exchange services, and image analytics, in order to be successful. Finally the chapter outlines briefly important characteristics of the PACS administrator and the evolving role of the medical imaging technologist as an informaticist.

9.1 Introduction

The previous chapters described data acquisition modalities for digital radiography and fluoroscopy, as well as digital mammography and digital tomosynthesis. These modalities generate a very large number of digital image files. For example, a CR examination consisting of two images with an image size of 2048 × 2048 × 12 will result in 16 MB of data. A digital mammography examination can now generate 160 MB of data. Additionally, the number of images generated in a multi-slice CT examination can range from 40 to 3000, and the image size is 512 × 512 × 12; then one examination can generate 20 MB of data and up [1, 2]. All digital X-ray imaging modalities as well as MRI generate very large amounts of digital images that result in what has been referred to as a data explosion. In 2004, Huang [3] points out that "the number of digital medical images captured per year in the US alone is over petabytes that is, 2^{15}, and is increasing rapidly every year." In 2012, in an article published in the *Indian Journal of Radiology and Imaging* [4], the authors report that in a PACS, depending on "the patient load, types of modalities, and the duration for which the images are to be stored, the storage size varies from terabytes (2^{12}) to petabytes (2^{15}) or even exabytes (2^{18}) and zetabytes (2^{21})."

These data files must be handled in such a manner that they can be easily displayed for viewing and interpretation, stored, and retrieved for retrospective analysis and be transmitted to remote locations within a healthcare facility, and to remote institutions, for the general management of the patient's medical condition. In the digital imaging environment, one of the central means of addressing this particular challenge requires the use of a picture archiving and communication system (PACS). PACS is now an important and significant tool in the digital radiology department that it is mandatory for technologists to have a firm understanding of the fundamental principles and technologies operating in a PACS environment. This is an essential element for effective and efficient use of the system.

The purpose of this chapter is to present a broad overview of the major components and technologies involved in PACS, as well as to lay the foundations needed for the study of the radiology informatics, the application of information technology (IT) to radiology.

9.2 PACS: A Definition

What is PACS exactly? Some individuals believe that perhaps it should be called IMACS (image management and communication systems); however, the more popular acronym is PACS. Throughout the years notable researchers in the field of digital medical imaging and those working in PACS have provided various definitions. For example the following are *definitions of PACS* in a chronological order:

- 2001: "*…are a collection of technologies used to carry out digital medical imaging. PACS are used to digitally acquire medical images from various modalities such as CT, MRI, Ultrasound (US), Nuclear Medicine (NM), and digital projection radiography. The image data and pertinent information are transmitted to other and possibly remote locations over network, where they can be displayed on computer workstations for softcopy viewing in multiple locations simultaneously. Data are secured and archived on digital media such as optical disks or tape and can be automatically retrieved as necessary*" [5].
- 2002: "*…PACS refer to a computer system that is used to capture, store, distribute, and then display medical images. For diagnostic imaging applications, PACS technology can be utilized to achieve near filmless operation*" [6].
- 2004: "*Picture Archiving and Communication System encompasses a broad range of*

technologies that enable digital radiology and the digital imaging department. At the most fundamental level, PACS represent the integration of image acquisition devices, display workstations, and storage systems-all connected via a computer network" [7].

- 2004: "*A picture archiving and communication system (PACS) is an inter-and intra-institutional computational system that manages the acquisition, transmission, storage, distribution, display, and interpretation of medical images. As such, the system is highly integrated with the imaging operation of the radiology department and with image-based clinical practice*" [8].
- 2004: "*A picture archiving and communication system consists of image and data acquisition, storage, and display subsystems integrated by digital networks and application software. It can be as simple as a film digitizer connected to a display workstation with a small image database or as complex as an enterprise image management system*" [3].
- 2012: "*A PACS is a system for the storage, transfer and display of radiological images.... and consists of a digital archive to store medical images, display workstations to permit physicians to view the images, and a computer network to transfer the images and related information between the imaging devices and the archive, and between the archive and the display workstations. There also must be a database program to track the location of images and related information in the archive, and software to permit interpreting, and perhaps referring physicians to select and manipulate images"* [2].

Basically, each of these definitions convey the same notion and meaning of PACS, that is, PACS is a popular term used to describe a digital medical imaging system or medical device where images are:

1. Acquired from the patient
2. Displayed on monitors for viewing and subsequent digital image post-processing
3. Stored and archived in electronic form
4. Transmitted to other locations via computer communication networks

The 2012 definition however identifies a database program and software to allow for interpretation and image manipulation. Each of these will be subsequently described later in the chapter.

9.3 Historical Development

The research leading to the development and implementation of PACS is quite involved and dates back several years. In 1970, for example, Dr. Paul Capp introduced the term "digital radiology." Later, in 1980, Heinz Lemke, in Berlin, introduced the notion of digital image communications. From the period 1982 to 2002, several conferences/meetings were presented, all dealing not only with the feasibility of the technology to accomplish filmless radiology but also the technical elements and economics of full implementation. Today, PACS continue to evolve at a rapid rate, and it is becoming more and more commonplace in radiology departments all over the world; however, another technology referred to as vendor neutral archive (VNA) has replaced the PACS in some instances.

One individual who has not only championed and worked in the development of PACS at the University of California at Los Angeles (UCLA) in 1991, and at the University of California at San Francisco (UCSF) in 1995, and who has become a renowned expert on PACS is Dr. H.K Huang, DSc. FRCR (Hon). He is currently a professor of Radiology and Biomedical Engineering at the University of Southern California, Los Angeles, chair professor of Medical Informatics at the Hong Kong Polytechnic University, and honorary professor of Shanghai Institute of Technical Physics, the Chinese Academy of Sciences. Dr. Huang has written extensively on PACS and imaging informatics [4] and has provided the radiology community with what is considered a seminal work dealing with the evolution and comprehensive description of the fundamentals of the PACS concept and serves as an essential tool for the medical imaging community.

Dr. Huang [3] has categorized PACS into five models based on how they are implemented. These are:

1. *The Home-Grown Model*: This is a PACS developed by the radiology department and hospital using technical components from various manufacturers to suit the needs of the hospital.
2. *The Two-Team Effort Model*: This model is based on collaboration of individuals from both outside and inside the hospital to address PACS specifications and a manufacturer to address implementation issues related to the PACS.
3. *The Turnkey Model*: In this model the PACS is developed by a manufacturer who sells it to a hospital and accepts full responsibility for implementing the system for use by technologists, radiologists, biomedical engineers, medical physicists, and other related personnel such as information technology (IT) staff.
4. *The Partnership Model*: In this model the manufacturer and the hospital work together to ensure the optimum performance and integrity of the system, via training for personnel, upgrading, and general system maintenance.
5. *The Application Service Provider (ASP) Model*: In this model "a system integrator provides all PACS-related service to a client, which can be the entire hospital or pa practice group. No IT requirements are required by the client" [3].

Early PACS were based on the use of the UNIX operating system for display and processing workstations. Furthermore, user interfaces posed several problems for the user, since they were not intuitive. The lack of PACS interfacing to information systems such as radiology information systems (RIS) and hospital information systems (HIS) resulted in workflow problems, such as the delivery of data to specific locations at specific times, a function carried out by technologists in a film-based imaging department.

Today, these problems have been solved; however, PACS are always in a state of evolution for the purpose of improving performance efficiency. For example, the development and refinement of enabling technologies, such as communication standards, computer and networking speeds, storage and archival devices, PACS workflow and image management technologies, and PACS/RIS/HIS integration, are ongoing activities. A very recent initiative is Integrating the Healthcare Enterprise (IHE). The purpose of IHE is to ensure that all digital imaging systems and information systems from different vendors can communicate with each other using communication standards to integrate all patient clinical information for the purpose of viewing such information collectively. IHE will be described later in this chapter.

Last, but certainly not least, we have seen the evolution of a "PACS administrator," a technologist who should be well-versed in the "language" of PACS to be able to meet the needs of the technologies required for a filmless imaging department. Perhaps, the PACS administrator and all radiologic technologists may have to be well educated in medical informatics and, in particular, radiology informatics (the application of IT to radiology operations).

9.4 PACS: Major Components and Core Technologies

The major components of a PACS and their functional relationships are shown in Fig. 9.1. These include *digital image acquisition modalities*, *database/image server, web server, archive server, short-term and long-term archive, and image display subsystems*, all connected via *computer networks*, using various gateways and switches. In order to extend its functionality and usefulness, the PACS is integrated with the RIS (and HIS as well). Furthermore, the system can serve to connect to remote locations via the Internet.

Communication and integration of all of these components require the use of communication protocol standards. In this regard, two data standards are now commonplace in a PACS environment:

1. The Digital Imaging and Communications in Medicine (DICOM) standard that essentially addresses the communication of images (image data).
2. Health Level-7 (HL-7) is a communication standard configured for information systems and addresses the communication of textual data, such as patient demographics, admission and discharge transfer, and the type of imaging examinations and radiology reports.

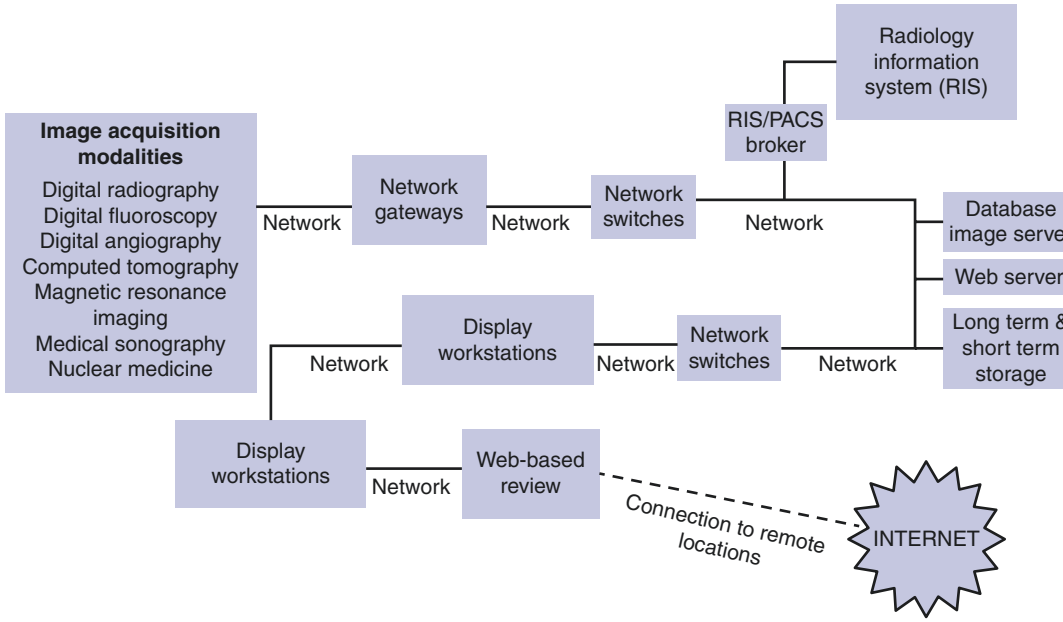

Fig. 9.1 The major components of PACS and their functional relationships

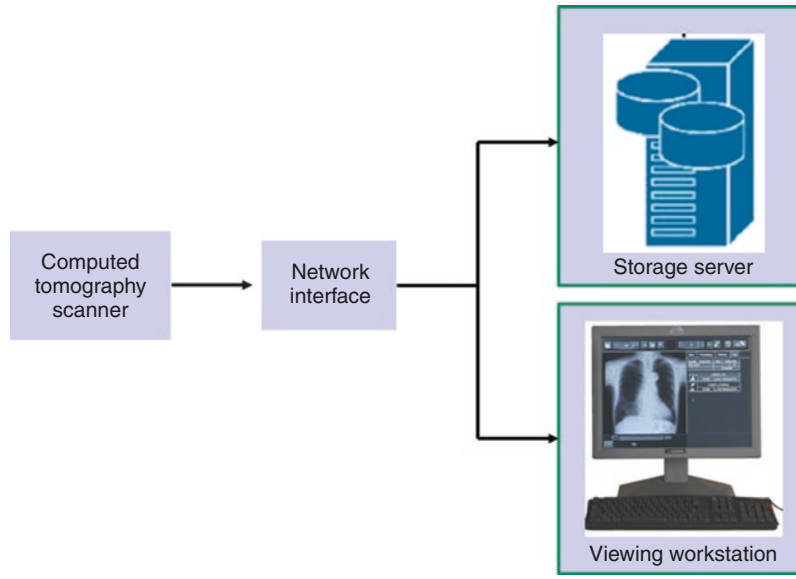

Fig. 9.2 A mini-PACS in which a single imaging modality (CT) is connected to a storage server and viewing workstations via a local area network

While it is not within the scope of this chapter to elaborate on these two data standards, the bare essentials of DICOM will be highlighted later.

PACS can be classified according to their size and scope. If a PACS is dedicated to a single digital imaging modality such as computed radiography (CR), flat-panel digital radiography (FPDR), computed tomography (CT), magnetic resonance imaging (MRI), nuclear medicine (NM), or diagnostic medical sonography (DMS), for example, it is usually called a mini-PACS, in which case a single local area network (LAN) is a central feature. This is illustrated in Fig. 9.2. On the other hand, PACS can be a large-scale system, when it includes all digital imaging modalities connected to a digital archive and display workstations and links to a

RIS/HIS using either an extended LAN or a wide area network (WAN), as is illustrated in Fig. 9.1.

9.4.1 Image Acquisition Modalities

To be a part of a PACS, *image acquisition modalities* must be digital in nature. These include CT, MRI, CR, FPDR, digital fluoroscopy (DF), digital mammography (DM), film digitizers, NM, and DMS. Digital image acquisition represents the first point where data and images are entered into a PACS, and, therefore, it is important that such entries are error-free since these mistakes will have negative consequences in the imaging chain.

Digital image acquisition modalities are of two types: those that are inherently digital such as CT and MRI [2], where the image data is obtained from the scanners at the full spatial resolution and grayscale (bit depth) compared to frame grabbling. Frame grabbers digitize the analog signals obtained from the image receptor, and images are sent to an image display device such as a cathode ray tube (CRT) monitor for viewing. An example of this is digital fluoroscopy where the output signal from the charged couple device (CCD) is digitized, processed, and subsequently sent to the display monitor for image viewing. The frame grabber grabs the same signal and converts it into a digital signal. In this case, image quality is limited in the grayscale resolution of the video signal [2]. Additionally, if the radiology department already has a large number of film images acquired with film-screen cassettes, it may be necessary to convert these images to digital data. One such device used for this purpose is the film digitizer. It is not within the scope of this chapter to describe the details of digital image acquisition modalities, including the film digitizer; however, it is important to highlight an important attribute of image acquisition relevant to PACS, and that is image data sets.

9.4.2 Computer Networks

As can be seen in Fig. 9.1, the hardware components of PACS and the associated information systems (RIS/HIS) are all connected. These connections are made possible via *computer networks*. Essentially, these networks allow information to be transferred and shared among computers. Network consists of both hardware components and the necessary software to enable the hardware to function. Networking is a complex topic, and therefore only a basic overview will be outlined in this chapter.

First, networks can be discussed in terms of *local area networks* (LANs) or *wide area networks* (WANs). The basis for this classification is the distance covered by the network. A LAN, for example, connects computers that are separated by short distances such as in a radiology department, a building, or two or more buildings. A WAN, on the other hand, connects computers that are separated by large distances, such as in another province or country. The *Internet* is a perfect example of a WAN. The network topologies for LANs include bus, star, or ring as shown in Fig. 9.3.

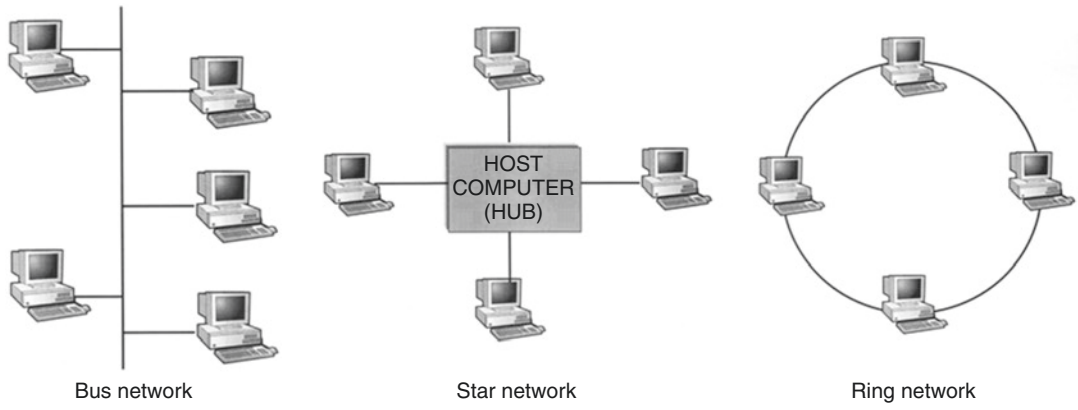

Fig. 9.3 Three network topologies for LANs (see text for further explanation)

In the bus configuration, computers (nodes) are connected via a single cable, as opposed to the star configuration, where all computers are connected to a central or host computer called a *hub*. Finally, in the ring topology, a computer is connected to two adjacent computers, and all computer connections form a ring.

In general, computers are connected by wires (coaxial cables, optical fiber cables), or they may be wireless connections (radio wave, microwave, and satellite links), capable of transferring data at different rates. This data transfer rate of the network is called the *bandwidth*. The bandwidth will vary depending on the physical connection between computers. The unit of bandwidth in this case is in megabits per second (1Mbps = 10^6 bps) or in gigabits per second (1 Gbps = 10^9 bps).

Information is sent through the network that is divided into packets of fixed or variable sizes and is sent via switching devices to the appropriate computer (or other switching devices) on the network. An individual device is referred to as a *node*, and the connections among nodes are called *links*. Other hardware components such as bridges and *routers* play an important role in sending the packets and ensuring that they get to the correct destination computer.

In order for information to be communicated through the network, *network protocols* are significant to this task. These are *communication protocols* that must be executed by the hardware and software. One such popular protocol in a PACS environment is the Transmission Control Protocol/Internet Protocol (TCP/IP) (actually two protocols, the TCP and the IP). TCP divides the information to be transferred into packets.

Networks often feature a dedicated computer (that is, of course, part of the network) capable of providing various services such as data storage *(file server)*, printing *(print server)*, e-mail *(e-mail server)*, and web access *(web server)*.

In the case of very large computer networks, LANs can be linked using devices referred to as bridges, the result of which generates an extended LAN. Additionally, routers are also used to connect larger and separate networks. The Internet (capital "I") is the largest network of networks. Finally, an *Intranet* is another type of network based on TCP/IP and used within a single organization, such as the radiology department. Another important network concept is long distance telecommunication links. These links are classified according to their data transfer rate, among other things.

The slowest and cheapest link, for example, is a telephone modem with transfer rates of 56 kilobytes/sec (kbps). Others include Integrated Services Digital Network (ISDN—128 kbps), digital subscriber line (DSL—up to about 1.5 Mbps), cable modems (500 kbps—10 Mbps), point-to-point links such as T1 (1.544 Mbps) and T3 (44.7 Mbps), and optical carriers (OC) that use optical fibers for data transmission (OC 1 = 51.84 Mbps; OC 3 = 155 Mbps [2]).

The transfer times for images to get to their destination is of vital importance to radiologists, and therefore the type of network technology and bandwidth used is an essential element that must be considered when implementing PACS. One final concept about networks relevant to this chapter is that of *network security*.

In a PACS/RIS/HIS environment, network security is paramount. One major security feature is a firewall. This is "a router, a computer, or even a small network containing routes and computers that is used to connect two networks to provide security" [2]. In addition, privacy, authentication, and integrity are vital security concerns surrounding an electronic radiology department. While privacy or confidentiality ensures access only to persons from whom the information is intended, "authentication permits the recipient of information to verify the identity of the sender and permits the sender to verify the identity of the recipient. Integrity means that the received information has not been altered either deliberately or accidentally" [2].

9.4.3 The PACS Main Computer

Images and data acquired from patients are sent via computer networks to the PACS main computer that is the heart of the system and consists of a "high-end" computer or server. This computer is sometimes referred to as the PACS

controller, the *database server* (Fig. 8.1) or the *image server*. Images and patient data (demographics, for example) are then sent from the acquisition gateway computer, the HIS and the RIS, to the PACS controller which has a database server as well as an archive system.

The PACS computer is well equipped with several central processing units that can process data very quickly and has a good deal of internal memory (both random-access memory *(RAM)* and cache memory) to assist in handling the vast amounts of data that is input into the computer.

9.4.4 Image Storage

The results of computer processing are information and digital images. The images are then sent through the computer network, to various computer workstations where they are generally displayed for viewing and stored temporarily and/or permanently for retrospective viewing and analysis.

As noted earlier, the digital images acquired in a total digital radiology department (filmless imaging) are large files with varying matrix sizes. For example, a typical CR image is about 2048 × 2048 matrix by 2 bytes per pixel, and a CT image is usually 512 × 512 matrix by 2 bytes per pixel. These two characteristics alone (matrix size and bit depth) place huge demands not only on storage requirements but also on the speed of transmission over the network.

Current storage technologies for PACS include solid-state memory (RAM), magnetic data carriers (disk and tapes), and optical disks. Each of these data carriers varies in terms of cost, storage capacity, and access time. While RAM and magnetic disks are expensive, optical and magnetic tape storage are inexpensive. Additionally, while the storage capacities for RAM and magnetic disks are limited, storage capacity is large for optical disks and magnetic tapes. For example, while RAM can be in the MB (1000 kbytes) to GB (1000 Mbytes) range, optical disks and magnetic tapes can store image files in the terabyte (TB—1000 Gbytes) to pentabyte (PB) (1PB = 1000 TB).

One popular storage technology used in PACS is the *redundant array of independent disks (RAIDs)* which can contain several magnetic or optical disks that can perform as a single large disk drive. The result is an automated library system (ALS) or a "jukebox." RAIDs are used primarily for short-term storage. In addition, robotic technology is now used to manage the movement of storage media to and from the storage shelves and image reader components.

Digital linear tape (DLT) is usually 0.5 in. magnetic tape that is intended for long-term storage of images. Multiple DLT drives can be arranged to create a "jukebox" similar to the RAID situation. It is important to note here that these image storage components can be connected by a sub-network called a storage area network (SAN), a fiber optic high-speed network (up to about 100 megabits/s) sometimes referred to as fiber channel. SAN has been designed solely for image storage as shown in Fig. 9.4.

Storage can be online, near-line, and off-line. While online storage cannot be removed and provides immediate access to images (e.g., RAM) within seconds, near-line storage is removable (e.g., RAIDS) with longer image retrieval times. Off-line storage on the other hand, refers to storage devices that must be retrieved by an individual and loaded into a drive for access to images. Off-line storage is of course cheap and provides the largest storage capacity.

9.4.5 Image Compression

One effective way to manage the size of image files for transmission and storage in medical imaging is that of *image compression* [9]. Image compression is a topic in itself that is quite complex and beyond the scope of this chapter; however, the following basic facts are noteworthy for radiologic technologists:

1. The purpose of compression is to speed up transmission of information (textual data and images) and to reduce storage requirements.

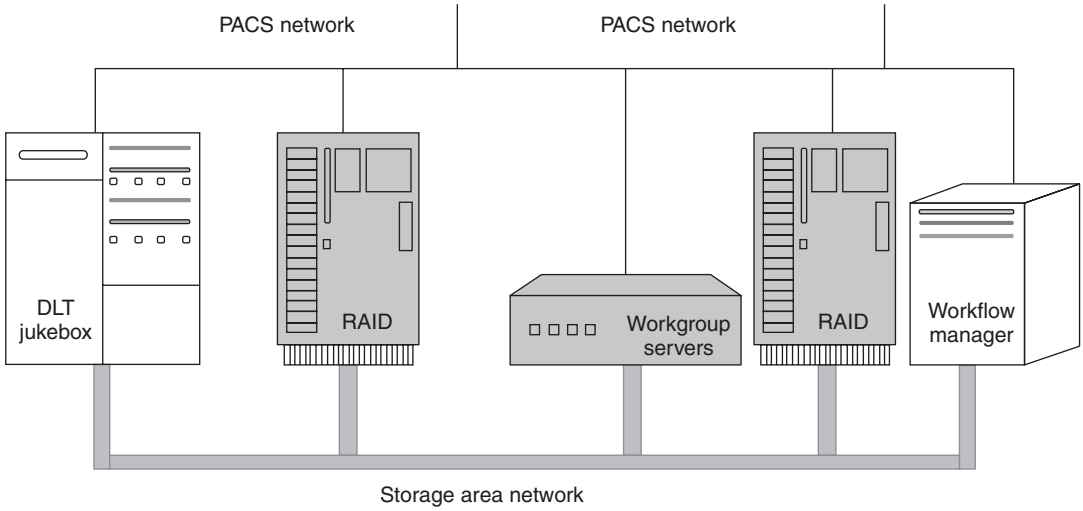

Fig. 9.4 A dedicated image storage network called a storage area network (SAN) for connecting image storage devices in PACS

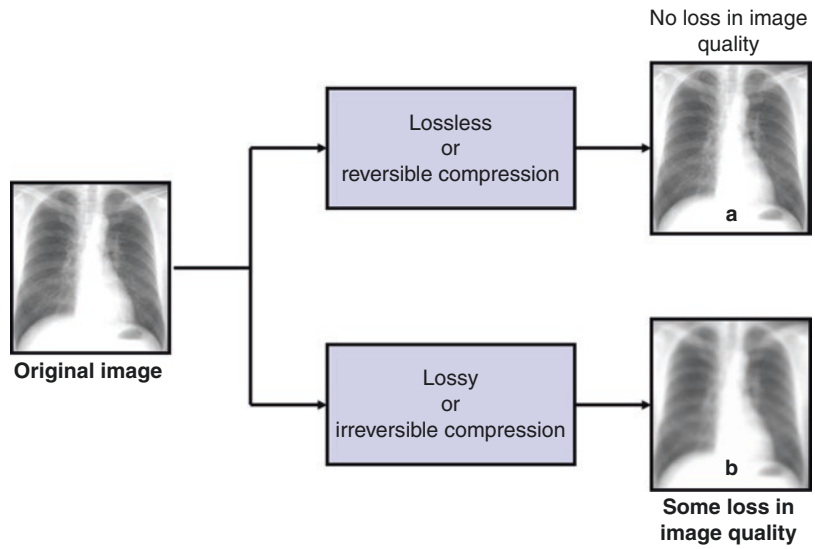

Fig. 9.5 The visual effect of two image compression methods used in PACS: lossless or reversible compression in which there is no loss in image quality (**a**) and lossy or irreversible compression (**b**) in which there is some loss in image quality. Higher compression ratios are possible with the latter

2. Several image compression methods are available, each providing advantages and disadvantages. Compression can be either:
   (a) *Lossless compression* or reversible compression, where no information is lost in the process, as shown in Fig. 9.5a
   (b) *Lossy compression* or irreversible compression, where some information is lost in the process, as shown in Fig. 9.5b

JPEG (Joint Photographic Experts Group) image compression, for example, features both lossy and lossless compression methods. For medical images, JPEG is DICOM's present compression method. Furthermore JPEG 2000 uses something referred to as wavelets (wavelet compression) in an effort to decrease the compressed image size and enhance image quality [9].

The *compression ratio* is the element that reduces the size of the image file, and therefore it is a very important variable in studies examining the effects of image compression on image quality. Huang [3] provides a brief explanation of compression ratio as follows:

> *The compression ratio between the original image and the compressed image file is the ratio between computer storage required to save the original image and that of the compressed data. Thus a 4:1 compression on a 512 × 8 = 2,097,152 bit image, requires only 524,288 bit storage, 25% of the original image storage required. There is another way to describe the degree of compression by using the term bit per pixel (bpp). Thus, if the original image is 8 bpp, a 4:1 compression means the compressed image becomes 2 bpp.*

The size of the compressed image is influenced by the compression ratio (see definition of terms), with lossless compression methods yielding ratios of 2:1 to 3:1 and lossy or irreversible compression having ratios ranging from 10:1 to 50:1 or more [3]. It is well known that as the compression ratio increases, less storage space is required and faster transmission speeds are possible, but at the expense of image quality degradation [9].

It is also well established that the quality of digital images plays an important role in helping the radiologist to provide an accurate diagnosis. At low compression ratios (8:1 or less), the loss of image quality is such that the image is still "visually acceptable" [9]. Liu et al. [9] reported that the recommendations for lossy compression ratios published by the Royal College of Radiologists for mammography, chest radiography, skeletal radiography, ultrasound, digital angiography, CT (all areas), and MRI are 20:1, 10:1, 10:1, 10:1, 10:1, 5:1, and 5:1, respectively.

9.4.6 Display and Analysis Workstations

A display and analysis workstation used in a PACS environment is a *computer workstation* consisting of hardware and software to facilitate the display of digital images for diagnostic interpretation and for review purposes. The PACS workstation is often referred to as a soft-copy display workstation.

Two types of image display formats are available for these workstations, a landscape display (Fig. 9.6a) and a portrait display (Fig. 9.6b). While the former displays more pixels in the horizontal than vertical direction, the latter displays more pixels in the vertical than in the horizontal direction.

The generic hardware components of a PACS workstation include a computer system consisting

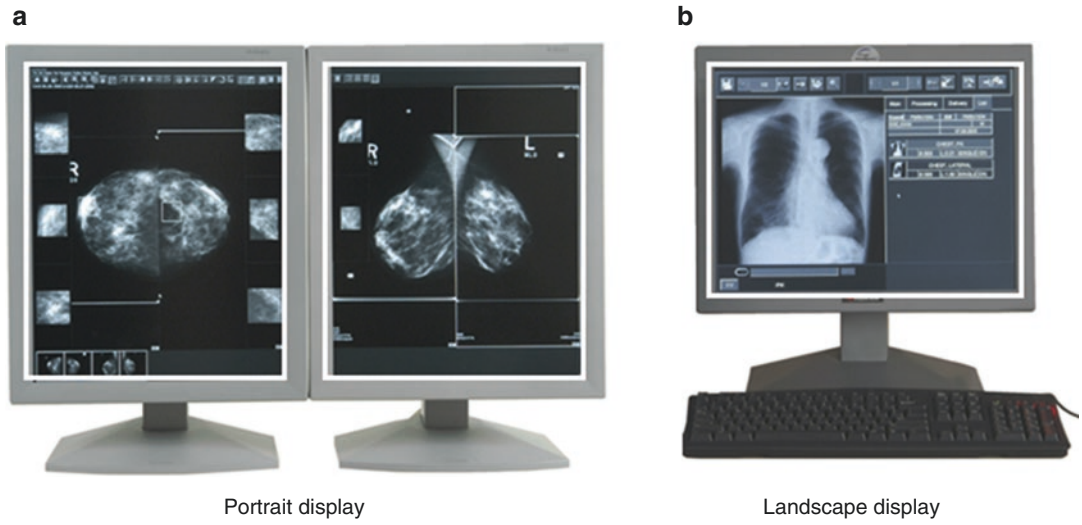

Fig. 9.6 Two types of image display formats for workstations used in a PACS environment, landscape display (**a**) and portrait display (**b**) (see text for further information)

of a central processing unit, RAM, network interface, serial controller, local disk storage, frame buffers, a display controller, and finally display device. Display devices can be a cathode ray tube (CRT) monitor or flat-panel displays. Flat-panel displays have recently become available for PACS and match the display requirements of CRTs; however, they are more expensive than CRTs.

The workstation is an essential component of PACS, since it provides a system-to-user interface. The ultimate goal of a PACS workstation is to provide the radiologist in particular, with a tool to make a diagnosis, and the technologist to assess the overall image quality before the image is sent to the PACS server. Therefore, the workstation must have adequate spatial and contrast resolution, as well as variable brightness and display speeds that facilitate diagnostic interpretation of digital images. It is important to note in this regard that, while the viewing task of the radiologist is lesion detection in particular, the viewing task of the technologist is overall image quality assessment to ensure that the image contains acceptable density and contrast and accurate positioning and is free of artifacts, all of which will have an impact on the diagnostic interpretation process.

In general, there are at least four types of workstations available in a PACS environment:

1. *High-resolution display workstations* (2.5K × 2.5K is not uncommon) that are used by radiologists for primary diagnosis, since their viewing tasks are fundamentally different than other physicians, medical physicists, and technologists. These are the most expensive of workstations. Higher-resolution monitors (e.g., 3K × 3K) are now becoming available.
2. *Medium-resolution display workstations* (1.5K × 1.0K is not uncommon) for secondary diagnosis, that is, review function that can be done by radiologists.
3. *Desktop workstations* for technologists and physicians, other than radiologists.
4. *Hard-copy workstations* for image printing. Image printing however may eventually become obsolete.

PACS workstations must also be fast enough to retrieve and display images from an archive or an imaging modality to meet the demands of radiologists. For example, it is important to get the workstation to display and image in at least 2 s [8].

Currently, multitasking workstations for PACS can actually let the user perform other tasks such as interpreting another study while images are being retrieved. Software for PACS workstations must be intuitive and versatile. Such software facilitates navigation functions image manipulation functions and other functions.

A navigation function is one that enables the user of the workstation to find the image or images to be reviewed. Several examples of navigation functions include work list, list all, compare, mark as read, folder display, image icons, consult, next patient (exam), and previous patient (exam), to mention only a few.

Image processing functions (described in Chap. 2) on the other hand allow users to adjust or manipulate the image to suit their viewing needs. Windowing (grayscale image manipulation) is the most commonly used image manipulation function. As illustrated in Fig. 9.7, the window width (WW = the range of the digital numbers making up the image) and the window level (WL = the center of the range of digital numbers) can be used to alter the image contrast and brightness, respectively. While increasing the WL (for a fixed WW) makes the image darker because more of the lower digital numbers are displayed, decreasing the WL (for a fixed WW) makes the image lighter, since more of the higher digital numbers are displayed [10].

Other image manipulation functions include outlining; boundary detection; region of interest (ROI) cleaning; grayscale invert; undo; pixel statistics; zoom and scroll; various image processing functions (edge enhancement, histogram modification); distance, area, and average gray-level measurements; and annotation. Equally important are image management functions such as delete, auto delete, print, redirect, scrapbook, and mark for teaching, to mention only a few.

Since PACS will be a part of the RIS/HIS environment, it is essential that workstations be

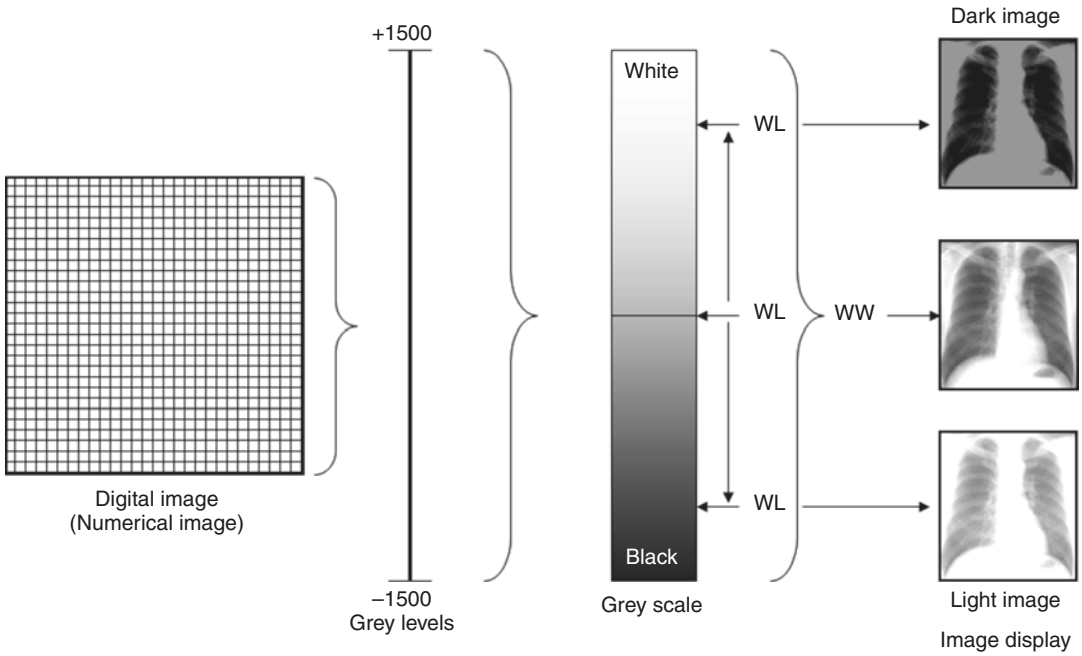

Fig. 9.7 Windowing can be used to manipulate the image brightness and contrast to suit the needs of the observer (see text for further explanation)

DICOM compliant. Furthermore, manufacturers should provide DICOM-conformance statements so that comparisons between devices can be made using the DICOM communication protocol as the basis. Finally, an important consideration surrounding PACS workstations and their use is that of workstation ergonomics. Ergonomics is the science and art of design that considers not only the physical aspects of the persons using the system but also their mental capabilities, in an effort to reduce the physical effort of those using the machine and mistakes that could be made as well.

Ergonomics also considers the design and use of furniture, heat, noise, and lighting, in an effort to maximize performance and minimize stress factors such as fatigue and anxiety.

9.4.7 The RIS/PACS Broker

A component of the PACS architecture shown in Fig. 9.1 is the *RIS/PACS broker*. While the PACS is concerned primarily with image data, the RIS deals with textual data such as patient demographics. In order for the PACS and the RIS to communicate with each other, such that patient information for Euclid is linked and associated with Euclid's images, a device is needed that would act as a translator so that communication can be effective and accurate. Such a device is referred to as the "*broker*" and, more specifically, the RIS-PACS broker. The integration of PACS with information systems (RIS and HIS) will be described subsequently.

9.4.8 The Web Server

The major purpose of the *Web Server* in PACS is to allow users to access images remotely, using Internet browser technology and microcomputers, that would allow access from within an institution or from outside the institution. Several strategies are currently available that provide distribution and display of images throughout the

enterprise. These will be described briefly later in this chapter under the heading called "Enterprise-Wide Image Distribution."

9.5 Workflow in a PACS Environment

In describing radiology operations, the term *workflow* is often used. This term as noted by Siegel and Reiner [11] refers to "the movement of patients, images, and information throughout the imaging department and healthcare enterprise." Technologists are all too familiar with workflow in a screen-film radiographic imaging (non-PACS environment) environment. In a PACS environment, workflow has an impact on clerical personnel, technologists, and radiologists. The ultimate goal of this workflow is to improve patient care and efficiency of various processes from patient registration and image acquisition to report generation, archiving, and image/report distribution throughout the healthcare enterprise. For a logical and smooth flow of images from acquisition to the time they are reported, stored, and distributed, some form of study management is needed. This is accomplished by what is referred to as the folder management function, provided by the PACS software.

9.6 PACS and Information Systems: Integration Overview

While PACS handles essentially images and is usually referred to as an image management and communication system, it requires other information about a patient (such as the patient demographics, clinical history, previous medical reports, and so forth), in order for the system to be effective and efficient in providing physicians' access to all pertinent information during the care and management of the patient. Such information is available from the HIS, RIS, and other information systems that are key components in a digital healthcare facility. Integration therefore serves to combine all the components into a whole system so that they can all communicate with one another.

A description of PACS is not complete without a discussion of integration of images and information systems. The discussion will focus briefly only on information systems pertinent to digital radiology operations.

9.6.1 Information Systems for Digital Radiology

An *information system* is a computer-based system that processes data to produce information that can be understood by people using the system in order to solve problems. Several authors, such as Bushberg [2], Huang [3], and Samei et al. [8], have discussed the role of information systems integration in radiology. They have identified at least five separate information systems in an electronic radiology department. These are:

1. PACS
2. RIS
3. HIS
4. Voice recognition system
5. Electronic teaching/research file system

Only the RIS and the HIS will be highlighted in this section.

A *HIS* is a computer-based information system used to gather not only medical information about a patient but also all activities related to the hospital's administration, such as billing, accounting, statistics, personnel, budgets, and material management, to mention only a few. There are usually at least two major components, the hospital business and administrative component and the hospital operation component [3]. These two are linked, and special software is used to allow the HIS to distribute HL-7 data to other systems external to the HIS.

A *RIS* is also a computer-based information system that could be a stand-alone system, or it may be integrated with a HIS. Some of the functions performed by the RIS are:

- Patient registration
- Exam scheduling
- Patient tracking
- Film archiving
- Report generation
- Administration and billing
- Documentation
- Inventory control
- Department statistics
- Communication standards

Explore the Internet for examples of commercially available RIS systems available.

9.6.2 Integration

The RIS/HIS/PACS must be able to exchange data and information in such a way that it is seamless to technologists, radiologists, and other related personnel in the hospital. Additionally the *Oxford Dictionary of Current English* states that to *integrate* means to "combine (parts) into a whole; complete by addition of parts." To perform this task requires a connection, an interface to facilitate this communication. This interface is a computer program that sets up the rules for communication so that different systems (RIS/HIS/PACS) can exchange data.

There are usually two types of RIS/HIS integration schemes that are available, one that uses a broker and the other that is brokerless and is usually the preferred method. The difference in these two integration schemes is illustrated in Fig. 9.8. *Integration* is an ongoing issue in a PACS environment as well as in the digital hospital, and currently more research is being done by information system vendors and different imaging modality vendors.

Integration also requires several elements, all of which play an important role in effective communication among systems. These include a clinical data repository, a data dictionary, mapping, a master patient index, uniform language efforts, and data exchange standards. Only the last one will be reviewed here.

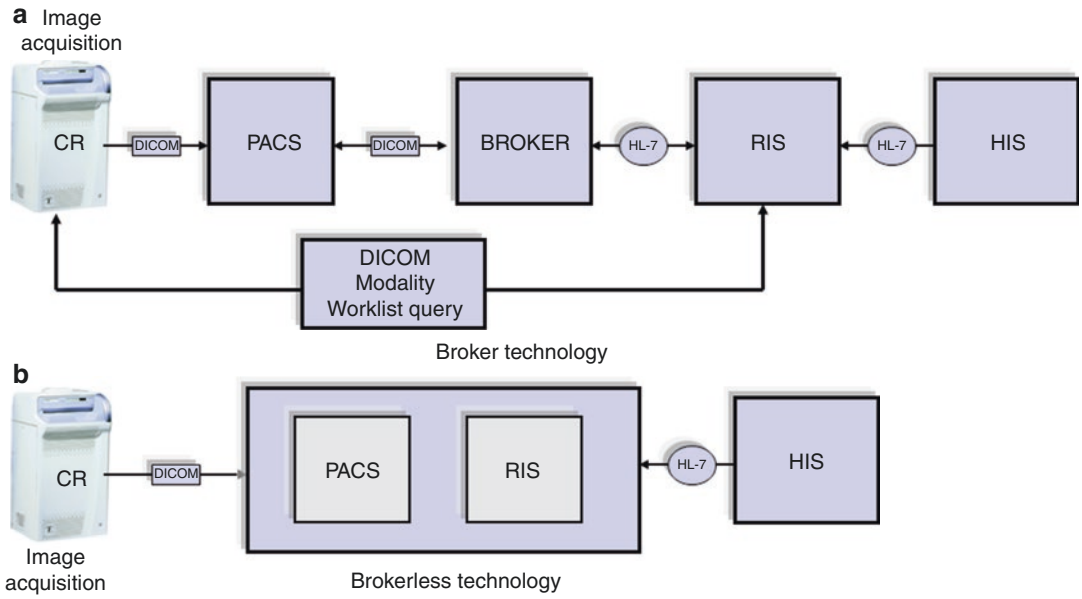

Fig. 9.8 Two types of RIS/HIS integration schemes are available, one that uses a broker (**a**) and the other that is brokerless (**b**). (**b**) is usually the preferred method

9.6.3 Integration or Communication Standards for PACS

The two information systems, the RIS and the HIS, must be integrated with the PACS and digital image acquisition modalities, for reasons mentioned earlier. In this regard several data exchange standards have been developed to provide this integration; however, two standards have become popular and commonplace, especially in the healthcare enterprise. These are *HL-7* (Health Level-7) and *DICOM* (Digital Imaging and Communications in Medicine). While HL-7 deals with the flow of textual information between the HIS and the RIS and other information systems, such as nursing information systems (NIS) in the hospital, DICOM deals primarily with the exchange of images in the radiology department and facilitates communication among manufacturer-specific imaging equipment.

In addition, for standards to become commonplace and useful requires the use of some sort of technical framework. Such a framework is that of the IHE (Integrating the Healthcare Enterprise). While HL-7 is beyond the scope of this book, DICOM and IHE will be explored briefly in this chapter.

9.7 DICOM®: The Bare Essentials

DICOM® is complex and details cannot be described in this overview section. For a more comprehensive coverage of DICOM, the student may explore the resources currently available on the Internet found at https://www.dicomstandard.org (active at the time of writing this chapter in January 2018). In brief, DICOM® is "the *international standard"* to transmit, store, retrieve, print, process, and display *medical imaging* information. It makes medical imaging information *interoperable* and *integrates* image acquisition devices, PACS, workstations, VNAs, and printers from different manufacturers and is actively developed and maintained to meet the *evolving technologies* and needs of medical imaging [12]. This site addresses topics such as About DICOM®, Standard, Activity, Using DICOM®, Resources, Conferences and News.

DICOM® was developed in a joint venture between the American College of Radiology (ACR) and the National Electrical Manufacturer's Association (NEMA) that in the early days was referred to as ACR/NEMA standard 2.0. DICOM® is the standard currently used in a PACS environment, simply because it offers:

- The specification of a network protocol that runs on top of the Internet standard protocol TCP/IP, allowing DICOM® devices to make use of commercial off-the-shelf (COTS) hardware and software
- The specification of strict requirement on the contents of the image "header" and the form of the pixel data itself for each type of modality, thereby improving interoperability
- The specification of a conformance mechanism, so that a user can decide whether or not devices are likely to interoperate
- An open standard development process that encourages the involvement and consensus of both vendors and users

What services does DICOM® offer for PACS? The answer to this question could be addressed in an entire course of DICOM®, since the standard is very large. DICOM is made up of 21 parts, with each part dealing with a specific process. For example, while Part 2 deals with conformance issues, Parts 3, 4, and 5 deal with information object definitions, service class specifications, data structures and encoding, and so on.

DICOM® basically specifies:

1. The type of communications referred to as *DICOM® service class*.
2. The types of data to be moved around the network and the data format. These are referred to as *DICOM® objects* (modalities). For example, a DICOM® object called CR is a data format for transmitting CR data. Other DICOM® objects include CT, MRI, NM, US, and so forth.

DICOM® services common to PACS include transfer and storage of images via the network,

query and retrieval of images, scheduling of acquisition, completion, notification, image printing, and transferring reports. These services are important to the transfer of image data among various devices in the PACS architecture.

In its most basic configuration, DICOM® communication can involve two devices, one called a *user* and the other a *provider*, hence the terms *service class user (SCU)* and *service class provider (SCP)*. The Internet, for example, provides us with a nice analog of this type of communication. When you use your computer to access the Internet from anywhere, you are a service user. Your request to access the Internet is answered by your Internet service provider (ISP).

In a simple configuration (Fig. 9.9), a CR unit sends images to a PACS for storage. In this case, the PACS is thought of as the provider (SCP), since it provides the storage for the CR images; and the CR unit is thought of as the user (SCU), since it uses the storage facility on the PACS, because both devices (CR unit and PACS storage) support the DICOM service class-store. Other configurations exist and can range from several imaging modalities coupled to printers and workstations to a large centralized or a large distributed PACS architecture.

Other DICOM® service classes are available. These include verification, modality work list management, performed procedure step, store, storage commit, print, and *query/retrieve*. For example, the function query/retrieve can be supported as an SCP and a SCU and can be applied to a number of situations. In a query/retrieve function, an image stored in the PACS archive (SCP) can be sent to a PACS display station (SCU).

Yet another example of query/retrieve function is a situation where the radiologist wants to access the PACS archive and display previous images of a patient on the CT console. These are possible because the modality and the PACS archive support query/retrieve SCU and SCP, respectively. In addition, DICOM® uses the term "role" to indicate that a device can be either a "user" (SCU) of a service or a "provider" of a service (SCP) as illustrated in Fig. 9.9.

Additionally, DICOM® specifies information object definitions (IODs) to address what is contained in images (image data and related information) from different imaging modalities. IODs are extremely useful in a PACS environment since images vary in size, acquisition parameters, and textual information.

Another element of DICOM® is that of a *Service-Object Pair (SOP) Class*, which includes both service class and an IOD. For an MRI scanner that sends only MRI data to the PACS, for example, the device specifications should show that the service class is store, the role (SCU, SCP) is SCU, and the SOP is MRI. Similarly, for a PACS archive that supports DICOM® store as a SCU, the device specifications should show that the service class is store, the role (SCU, SCP) is SCP, and the SOP would be CT, MRI, NM, CR, US, etc. Last but not the least (since this discussion can be extended to an entire course on DICOM – we have just touched the surface) is the notion of *DICOM® conformance statements*. These are statements that deal with specific ways of implementation using the service

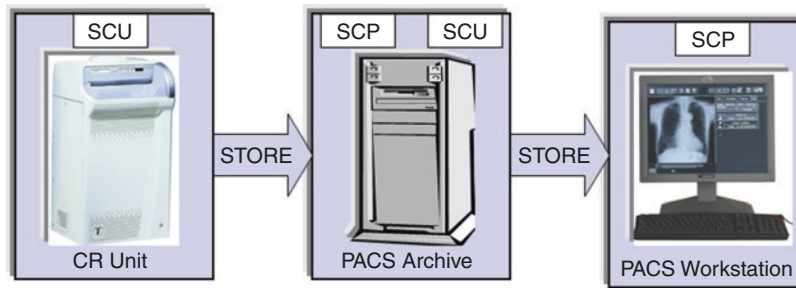

Fig. 9.9 DICOM communication can involve two devices, one called a *user* and the other a *provider*, hence the terms *service class user* (SCU) and *service class provider* (SCP). In this basic configuration, a CR unit sends images to a PACS for storage (see text for further explanation)

classes, information objects, and communication protocols that are supported. DICOM® conformance is seen as a useful specification and implementation strategy.

9.8 Integrating the Healthcare Enterprise: A Brief Overview

9.8.1 Problems with DICOM and HL-7

The communication standards DICOM and HL-7 do not address message sequencing and problems that may relate to connectivity of various equipment from different vendors. To solve these problems, *Integrating the Healthcare Enterprise (IHE)* enters the picture.

9.8.2 IHE Process Flowchart

IHE is "an initiative by healthcare professionals and industry to improve the way computer systems in healthcare share information. IHE promotes the coordinated use of established standards such as DICOM and HL7 to address specific clinical needs in support of optimal patient care. Systems developed in accordance with IHE communicate with one another better are easier to implement and enable care providers to use information more effectively" [13].

The major role of IHE is to facilitate communications between various computer-based healthcare information systems and imaging modalities and PACS vendors through the integrated use of DICOM® which handles mainly image data and HL-7 which handles textual data exchange. This effort will serve to increase the pace of the digital healthcare enterprise integration and to remove data redundancy and task repetition by narrowing the information loop between information systems (e.g., RIS/HIS) and imaging devices (PACS and image acquisition modalities). It will allow the radiologist, for example, to view relevant patient history, laboratory data, clinical history, and so on while viewing images at the display monitor for diagnostic interpretation. To accomplish these functions, IHE provides several *process flowcharts* that illustrate resources and tools as shown in Fig. 9.10 such as standards, technical specifications, integration profiles, and technical frame-

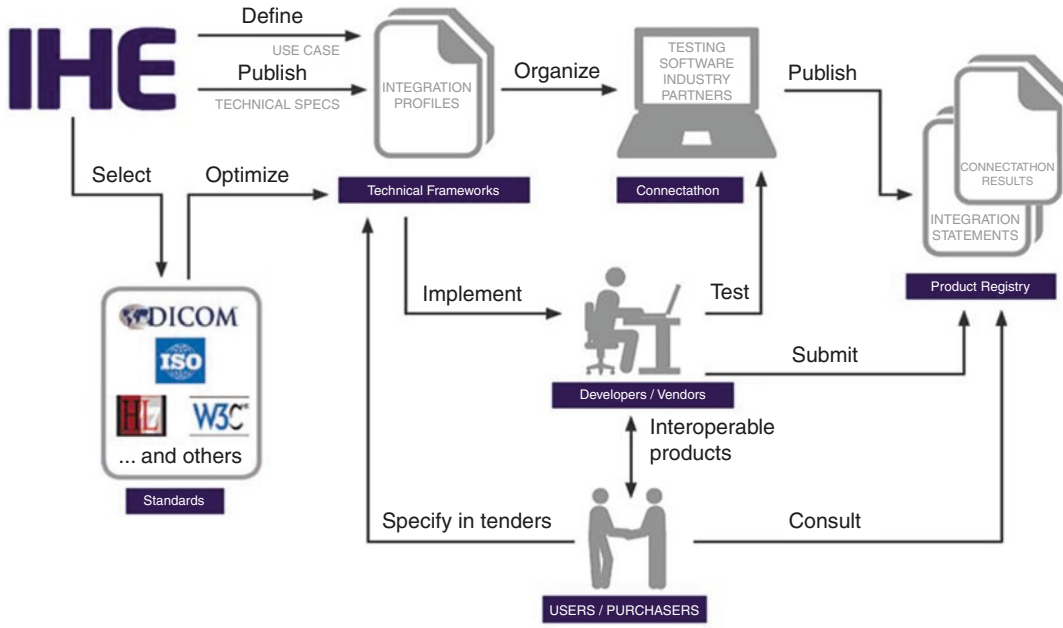

Fig. 9.10 The *process flowchart* of the IHE that illustrates resources and tools such as standards, technical specifications, integration profiles, and technical frameworks (Used by permission of the IHE)

works, for the purpose of helping users not only to integrate systems but also to share information in an efficient and effective manner. Resources can be found at https://www.ihe.net/About_IHE/.

A *technical framework*, for example, is built around at least three major components, namely, a data model, actors, and integration profiles. While the data model is based on communication standards DICOM® and HL-7, actors are defined by IHE as "information systems or components of information systems that produce, manage, or act on information associated with operational activities in the enterprise" [13, 14]. Several actors and their roles are outlined by the IHE and include the patient registration, image acquisition modality, image archive and display systems, image manager, master patient index (MPI), report creator, report manager, print server, post processing manager, report repository, system scheduler, and order filler, to mention only a few.

The IHE also defines integration profiles, which represent a central feature of IHE specifications. There are several integration profiles such as scheduled workflow, patient information reconciliation, post-processing workflow, charge posting, presentation of grouped procedures, consistent presentation of images, key image notes, simple image and numeric reports, access to radiology information, reporting workflow, evidence documents, and basic security.

The primary function of the IHE integration profiles is to outline the roles of actors in the DICOM and HL-7 environment. For example, the scheduled workflow profile (often viewed as the most important profile) ensures effective communication from the RIS to the imaging modality (via HL-7) which in turn captures the information sent from the RIS using DICOM and subsequently to the PACS, where images from the modality are sent for storage and archiving. The essential point of this example is that if an institution has imaging modalities, PACS and RIS from different vendors, IHE scheduled workflow profile ensures that the information exchange (communication) among these different vendor systems is seamless. Additionally, departments purchasing digital imaging equipment, PACS, and RIS systems should ensure that the vendors comply with the IHE profiles.

9.9 Enterprise-Wide Image Distribution and Viewing

In a PACS environment, it is important to deliver diagnostic reports to referring physicians and specialists, in a timely fashion. In the past, physicians relied on the legacy system (a system whereby a physician had to go to the radiology department to view films and read reports); however, this classic strategy has been replaced by the remote access concept or what has been typically referred to as "enterprise-wide image distribution" [8]. Remote access technologies used at that time included an extension of the traditional PACS approach where images can be sent to all locations in the hospital using the same network, server, and workstation infrastructure, thin client (web-based) approaches, and compact disk-read-only memory (CD-ROM) which does not require computer networks.

The growth of the *electronic health record* (EHR) in hospitals resulted in a wide range of clinical users with varying needs such as reviewing the different types of images within the EHR. One such fundamental need is image viewing, which now requires "an enterprise image viewer," a technology which is receiving increasing attention in the marketplace [15]. In a recent paper by Roth et al. [15], the authors define an *enterprise viewer* as "a thin-client or zero-client application used on any off-the-shelf device to distribute, display, and manipulate multispecialty image, video, audio, and scanned documents stored in separate centralized archives through, or standalone from, the EHR" [15].

Essential features of an enterprise viewer for the user are its toolsets. These toolsets include basic toolsets, advanced toolsets, specialty toolsets, and workflow toolsets. While a basic toolset includes navigation tools which enables the user to pan, zoom, rotate, measure angles and distances, as well as change the contrast and brightness of images through windowing (Chap. 2), advanced toolsets consist of several features for more advanced image processing, such as surface rendering, volume rendering, multimodality fusion, multi-planar reconstruction (MPR), bone removal, and maximal/minimal intensity projection (MIP/MinIP) [15]. Specialty toolsets on the

other hand features tools that are specific to other vocations such as cardiology, orthopedics, ophthalmology, and pathology [15]. Finally, Roth et al. [15] state that "enterprise viewers serving multiple specialties and use cases have gained favor for many reasons. These reasons include institutional pressures favoring scalable infrastructure, the business need presented by electronic health record image storage and viewing, data flow standard advances, viewer toolset improvements, and others" [15].

9.10 PACS in an Educational Institution

In the management and care of patients, PACS have been implemented largely in hospitals and clinics where healthcare and medical information are the primary focus. These systems can also be used for teaching and research purposes as well. In an environment for the education of medical radiation technologists such as radiological technologists, CT and MRI technologists, ultrasound technologists, radiation therapists, and nuclear medicine technologists, a PACS can prove to be an extremely useful tool for teaching and learning. More importantly, it can serve to prepare students for the interaction with PACS in the hospital.

A potential architecture for the use of a PACS for teaching and learning in medical imaging programs is illustrated in Fig. 9.11. The different types of activities that can be accommodated by such a system are:

1. Students (working in the laboratory of the educational institution) acquire images of phantoms using digital imaging modalities such as CR for example.
2. These images are assessed for image quality and are then sent to the PACS server and stored for retrospective analysis based on assignments from other courses. For example,

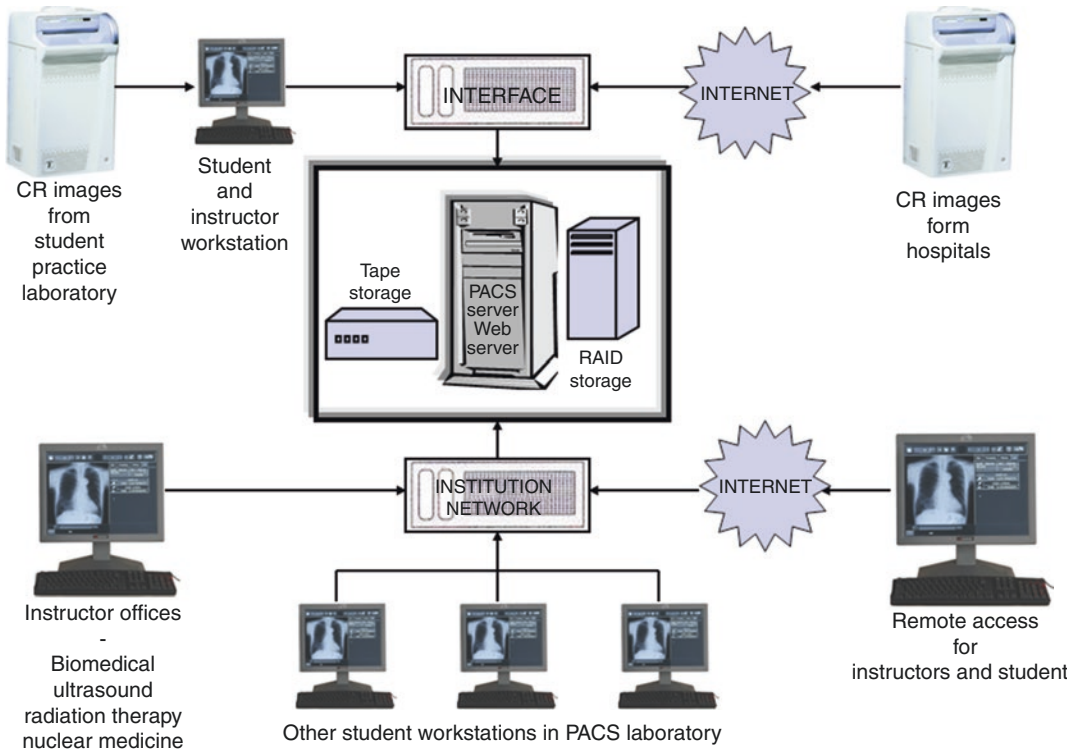

Fig. 9.11 A suggested framework for the use of a PACS for learning and teaching in several programs such as medical imaging, radiation therapy, and biomedical engineering technology

in a QC course, the instructor can assign a project whereby the student must access the QC image taken earlier in the laboratory exercise and evaluate it to meet the needs of the QC assignment.
3. An instructor can access student images for the purposes of assessment of position and techniques requirements covered in a laboratory exercise and/or a theory course assignment.
4. All instructors have access to the PACS server and hence can create teaching files for their respective courses.
5. Internet access to the PACS is possible for both instructors and students. Assignments based on images recorded in a laboratory exercise or images discussed in class can now be done at home.
6. Another important feature of the PACS scheme shown in Fig. 9.11 is that images can be imported form the affiliated hospitals and be housed in the teaching institution's PACS, for teaching/learning purposes.

The use of PACS for teaching and learning is an exciting venture and opens up a whole new dimension for both teachers and students.

9.11 PACS and Regulatory Approval

PACS is considered a medical device, and as such it is subject to federal and other regulations. In the United States, for example, the Food and Drug Administration (FDA) and the Health Insurance Portability and Accountability Act (HIPAA) govern such regulations.

9.11.1 Food and Drug Administration

The *FDA* notes that medical devices are used for the diagnosis, treatment, and prevention of disease in humans and animals and range from simple devices such as a tongue depressor and a bedpan to more complex devices such as lasers, devices based on microchip technology (pacemakers), and radiation emitting devices. Radiation emitting devices include X-ray equipment, ultrasound, radiation therapy machines, and so forth.

In the early days of PACS development, FDA approval was not required since PACS was considered accessory equipment to image acquisition modalities. However, with the evolution of more and more sophisticated PACS, the FDA now considers PACS as one of the five new classifications for medical image management devices. These are medical devices for image storage, image communications, image digitizer, hard-copy images, and PACS.

9.11.2 Health Insurance Portability and Accountability Act

The *Health Insurance Portability and Accountability Act (HIPPA)* became law in the United States in 1996, in an effort to reform healthcare by addressing several objectives relating to the protection of the patient's health information, standards for such information, fraud and abuse of such information, and so forth. The act essentially deals with the use and disclosure of such information. Interested readers should refer to the website at https://www.hhs.gov/hipaa/for-professionals/index.html for further details [16].

Since PACS stores, archives, and most importantly communicates information with RIS and HIS and to physicians inside and outside the hospital, this act is viewed as one intended to protect such information and essentially places a good deal of importance on the security of electronic data and information. If there is a need to share this information with related parties, then patient permission is required.

9.12 Vendor Neutral Archive in a PACS Environment

One of the fundamental problems with PACS is that not all PACS is DICOM conformant. PACS vendors have propriety approaches (conformance statements) "which may or may not conform to all that is expected of them and there may be a

few gaps and inconsistencies" [4]. These approaches allow for specific ways to store data and images. Because of vendor-specific conformance statements (which occur occasionally), it is usually difficult for an imaging department to migrate data from their "old" PACS to a new PACS. This problem can be solved with what is referred to as a *vendor neutral archive* (VNA).

A VNA is essentially an archive that "can be easily migrated, ported to interface with another vendor's viewing, acquisition, and workflow engine to manage medical images and related information" [4]. VNAs can operate on data from a vendor's PACS using a technical feature referred to as an application engine. While the position of a conventional DICOM PACS storage is shown in Fig. 9.12a, a VNA is located between the acquisition modalities and the PACS as shown in Fig. 9.12b. It is clear that the data are "pushed directly" from the imaging modalities and the data are sent to the PACS [4]. VNAs now function as an archive for the entire digital imaging enterprise as shown in Fig. 9.13. Furthermore, two key concepts of VNAs that are important are "neutrality" and "interoperability." While the former simply means that data must be stored in nonproprietary and interchange formats, the latter means "....the ability of different information technology systems and software applications to communicate, exchange data, and use the information that has been exchanged" [17]. In summary a VNA is an archive in which images and texts including related clinical information are stored in standard formats using standard interfaces to allow access to these data (without a particular vendor's control) by other systems.

The interested reader should refer to the article by Agarwal and Sanjeev [4] for more detailed account of VNAs. The authors outline such topics as archiving and its advantages, features of VNA, post-implementation issues, differences between DICOM archive and VNA, and the future of archiving with VNA. For example, as noted by Agarwal and Sanjeev [4], "VNA is able to perform "context management." This term denotes the ability of VNAs to present data in a format that is different from the original stored format. DICOM archive does not interface with RIS and/or HIS, whereas VNA can be interfaced with RIS and/or HIS. DICOM archive stores only DICOM objects, whereas VNA stores DICOM objects as

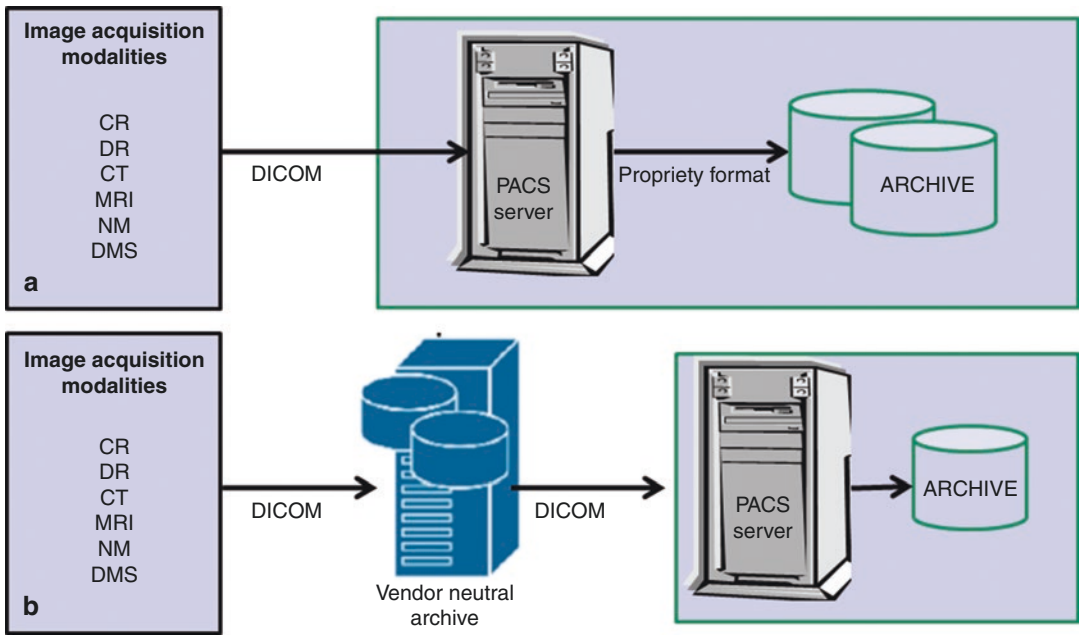

Fig. 9.12 While the position of a conventional DICOM PACS storage is shown in (**a**), *a VNA* is located between the acquisition modalities and the PACS as shown in (**b**)

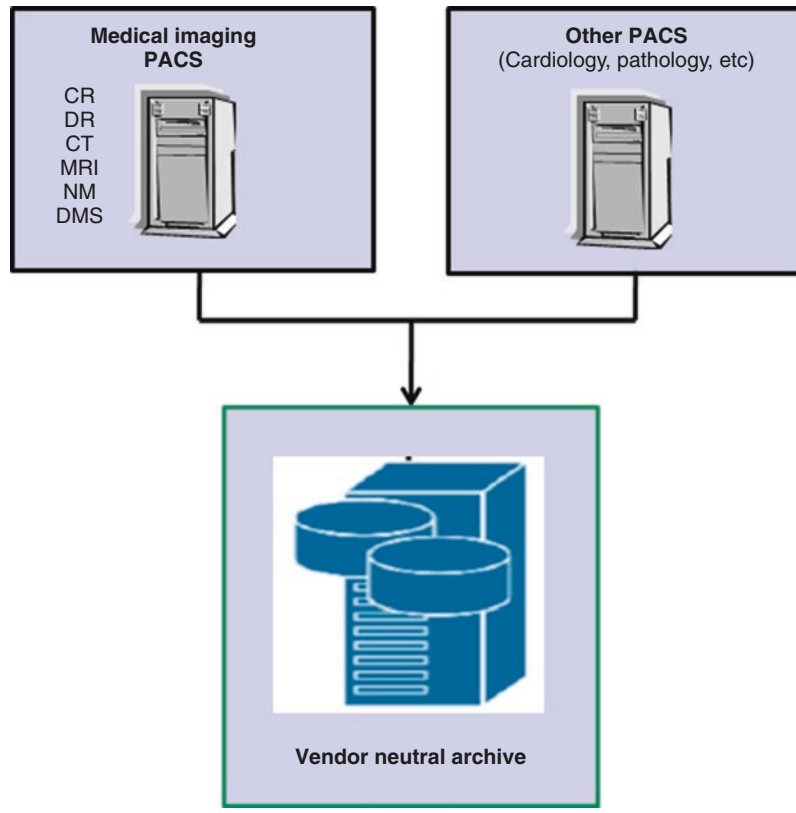

Fig. 9.13 The function of VNAs as an archive for the entire digital imaging enterprise

well as non-DICOM objects such as from .pdf, scanned documents, etc." [4]

9.13 Enterprise Imaging

The use of VNAs and PACS, including RIS to essentially handle all of its capture, viewing, storage, analysis, and communications of text and images produced by the various imaging modalities in a department, plays an effective role in image and information system management. The above brief overviews of PACS and VNAs illustrate that these technologies are still currently being used worldwide in the healthcare industry. A more recent approach to data archiving gaining popularity in healthcare institutions is that of *enterprise imaging*, a technology that offers improved efficiencies in data archiving and management. Enterprise imaging creates a comprehensive program that can combine data from different departments into one system. This topic has received increasing attention in the medical imaging community. Since there is a lack of information on this subject, the working group of the Healthcare Information and Management Systems Society (HIMSS) and the Society for Imaging Informatics in Medicine (SIIM) have published in the *Journal of Digital Imaging*, several papers [15, 18–21] that provide general overviews on major elements of enterprise imaging. One of the major purposes of these papers is to educate the community at large.

9.13.1 Definition

A formal definition of enterprise imaging offered by the working group of the Healthcare Information and Management Systems Society (HIMSS) and the Society for Imaging Informatics in Medicine (SIIM) has been developed for the medical imaging community. They state that enterprise imaging is "a set of strategies, initiatives and workflows implemented across a healthcare enterprise to consistently and optimally

capture, index, manage, store, distribute, view, exchange, and analyze all clinical imaging and multimedia content to enhance the electronic health record" [18].

9.13.2 Major Elements of an Enterprise Imaging System

According the HIMSS/SIIM workgroup, an enterprise imaging system that would be considered successful should consist of the following elements: governance, enterprise imaging strategy, enterprise imaging platform (infrastructure), clinical images and multimedia content, electronic health record (EHR) enterprise viewer, image exchange services, and image analytics. It is not within the scope of this chapter to outline details of these elements, and therefore the interested reader should refer to the paper by Roth et al. [18] for the more foundational descriptions; however, only three elements will be highlighted below, that is, the enterprise imaging platform, the clinical imaging and multimedia content, and the EHR enterprise image viewer.

Governance has been defined by the HIMSS-SIIM collaborative workgroup as "the decision-making body, framework, and process to oversee and develop strategies for the enterprise imaging program, technology, information, clinical use, and available financial resources." [15] The *enterprise imaging platform* is a key element of enterprise imaging that addresses infrastructure, modalities, devices, and integration and is characterized by several components as shown in Fig. 9.14. These include the EHR, the enterprise

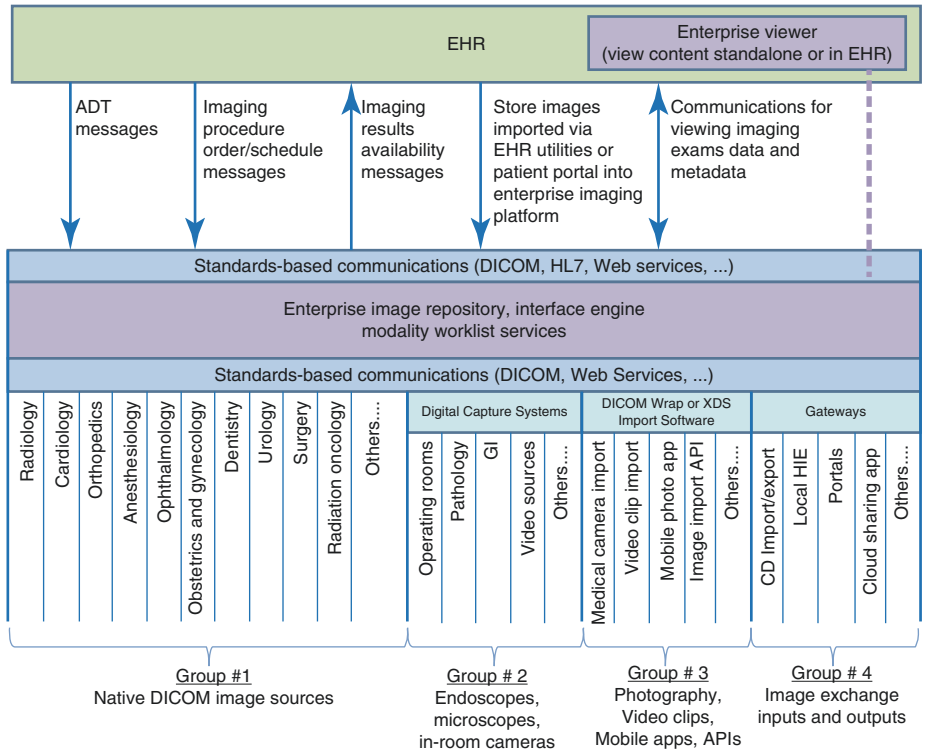

Fig. 9.14 An enterprise imaging platform provides the standard-based, enterprise infrastructure to support departmental imaging workflows. This includes modality worklist services, image archival, index, enterprise viewer application viewing within or outside the EHR, query/retrieve of imaging content from most departments, as well as image exchange capabilities [From Roth CJ, Lannum LM, and Persons KR. A Foundation for Enterprise Imaging: HIMSS-SIIM Collaborative White Paper. *J Digit Imaging*, 2016; 29:530–538. This article is distributed under the terms of the Creative Commons Attribution 4.0 International License (http://creativecommons.org/licenses/by/4.0/)]

image viewer, standard-based communications such as DICOM and HL-7, the enterprise image repository, and interface engine and modality worklist services. These form the core of the enterprise imaging strategies. Furthermore, a broad spectrum of enterprise imaging content is shown in Fig. 9.15 and includes not only diagnostic imaging but procedural imaging, evidence imaging, and image-based clinical reports. The *enterprise imaging viewer* was defined and described briefly in Sect. 9 above. In summary, the viewer must meet the needs (such as image review and image manipulation) of not only diagnostic imaging but also those of surgical subspecialties and other users such as cardiology, dermatology, mobile device users, ophthalmology, pathology, radiology, and research. For more details of the characteristics of enterprising imaging, the interested reader should refer to the seminal papers [15, 18–21] listed in the reference section.

9.14 The Radiologic Technologist as Informaticist: An Evolving Role

The rapid evolution and growth in information systems and technology in healthcare (RIS/HIS), PAC, and digital imaging modalities (filmless imaging) and notions of enterprise-wide image distribution, ASPs and the IHE, and more recently VNAs and enterprise imaging, will have an impact on the future role of the technologist in medical imaging. One such emerging role is that of "informaticist" or "informatician." Both terms have been derived from the popular term *information technology (IT)*. IT is defined as "the technology involving the development, maintenance, and use of computer systems, software, and networks for the processing and distribution of data" [22].

The definition reflects the nature of PACS/RIS/HIS and demonstrates the need for the tech-

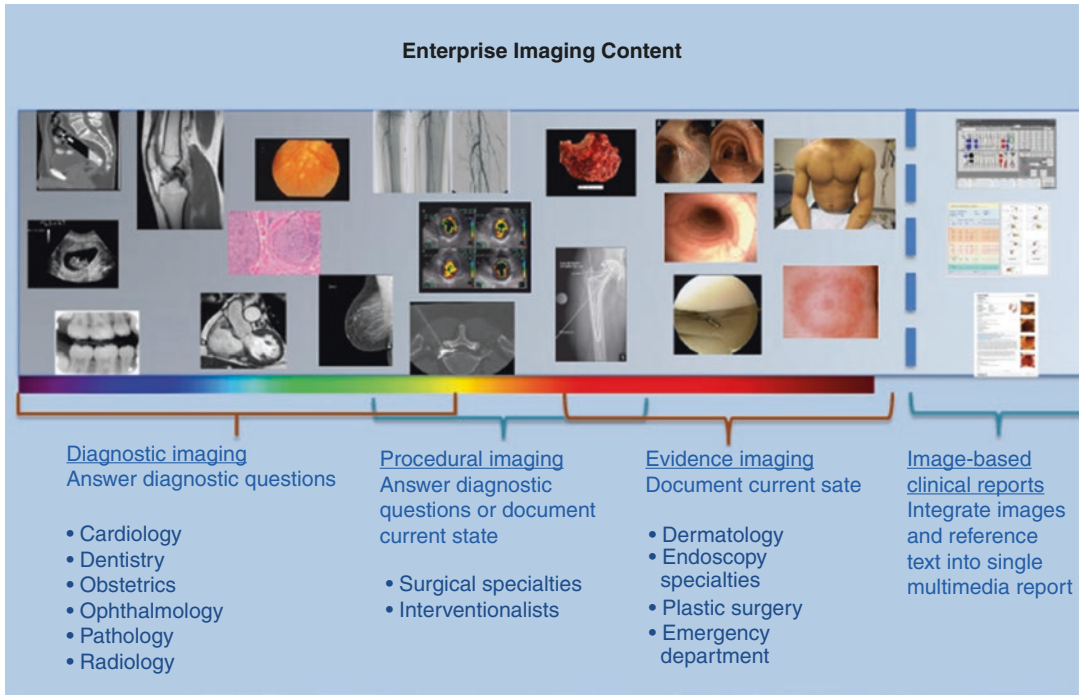

Fig. 9.15 The broad spectrum of enterprise imaging content and common use cases [From Roth CJ, Lannum LM, and Persons KR. A Foundation for Enterprise Imaging: HIMSS-SIIM Collaborative White Paper. *J Digit Imaging*, 2016; 29:530–538. This article is distributed under the terms of the Creative Commons Attribution 4.0 International License (http://creativecommons.org/licenses/by/4.0/)]

nologist working in a digital radiology department (filmless imaging) to be IT literate, that is, to be an informaticist or informatician. Already, we see the technologist assuming the role of "PACS administrator," a position that demands a good deal of understanding of computer and communication technologies.

9.14.1 PACS Administrator

Digital image acquisition modalities coupled to PACS and enterprise-wide image distribution systems have created the need for personnel trained in IT and computer technology, to ensure that the integrity of these digital imaging and computer communication systems, so as not to comprise patient care and management, image quality, and administrative functions. To address this need, radiology departments have created a new job title called the *PACS administrator*.

The PACS administrator is usually a senior technologist with some background and interest in computers and IT, or it may be an IT individual from the hospital IT department who provides support to the radiology department and works closely with radiology personnel to maintain system integrity.

The duties of a PACS administrator will be tailored to the needs of individual departments depending on the variety of digital image acquisition systems and PACS. The tasks or duties fall into three major categories: project management, system maintenance, and image and information management. A few examples of specific tasks would be:

1. Ensure that the PACS is working according to the needs of the department and the system specifications.
2. Check that workstations are functioning properly with optimum display of image brightness and contrast.
3. Assign and update passwords for staff.
4. Set up standards for using the system.
5. Maintain effective communications with other departments using the PACS.
6. Communicate with the facility's information systems (IS) or IT department.
7. Provide ongoing education and training for system users.
8. Liaise with PACS vendors and relate new developments to the facility's administrators and managers.
9. Workflow mapping.
10. Quality control of system components and so forth.

9.14.2 PACS Administrator Professional Certification

The PACS administrator role has become commonplace in digital radiology departments that already efforts are being made to provide certification for individuals interested in assuming this position. One such major effort is that of the PACS Administrators Registry and Certification Association (PARCA). This independent organization plays an active role in not only developing the skill set for PACS administrators but also providing access to certification examinations.

Skill sets are established by an advisory group with members who have extensive experience and knowledge in PACS and imaging informatics. PARCA offers certificates along two distinct paths. While one path is intended for individuals with a clinical background (e.g., radiological technologist), the other path is for individuals with an IT knowledge base. Three types of certification are offered: the Certified PACS Interface Analyst (CPIA), the Certified PACS System Analyst (CPSA), and the Certified PACS System Manager (CPSM). For further details, the interested reader should refer to the PARCA website located at http://www.pacsadmin.org/ (accessed Feb 2018).

9.14.3 Radiology Informatics Curriculum

In a total digital imaging and digital healthcare environments, it will become increasingly important that the education and training of radiologic technologists focus on curriculum that includes medical informatics and radiology informatics.

Specifically there should be courses dealing with digital image acquisition technologies, digital image processing, computer hardware and software, networking and connectivity, storage and display technologies, teleradiology, and telemedicine. In other words programs will have to prepare students to also assume the role of technologist as informaticist. This means that educators will also have to assume increasing roles in teaching radiology or medical imaging informatics. Radiology informatics or medical imaging informatics will be described in Chap. 10.

References

1. Bushong S. Radiologic science for technologists. 11th ed. St Louis: Elsevier-Mosby; 2017.
2. Bushberg JT, Seibert JA, Leidholdt EM Jr, Boone JM. The essential physics of medical imaging. 3rd ed. Philadelphia: Lippincott Williams and Wilkins; 2012.
3. Huang HK. PACS and imaging informatics. Hoboken, NJ: Wiley; 2004.
4. Huang HK. PACS-based multimedia imaging informatics: basic principles and operations. Hoboken, NJ: Wiley; 2018.
5. Agarwal TK, Sanjeev. Vendor neutral archive in PACS. Indian J Radiol Imaging. 2012;22(4):242–245.
6. Arenson RL, Andriole K, et al. Computers in imaging and healthcare: now and in the future. J Digital Imaging. 2000;13(4):145–56.
7. Siegel E, Reiner B. Image workflow. In: Dreyer KJ, Mehta A, Thrall JH, editors. PACS: a guide to the digital revolution. New York: Springer-Verlag; 2002.
8. Hori SC. In: Dreyer KJ, Mehta A, Thrall JH, editors. PACS: a guide to the digital revolution. New York: Springer-Verlag; 2004.
9. Samei E, Seibert JA, Andriole K, Badano A, Crawford J, Reiner B, Flynn MJ, Chang P. AAPM/RSNA tutorial on equipment selection: pacs equipment overview general guidelines for purchasing and acceptance testing of PACS equipment. Radiographics. 2004;24:313–34.
10. Liu F, Hernandez-Cabronero M, Sanchez V, Marcellin MW, Bilgin A. The current role of image compression standards in medical imaging. Information. 2017;8:131–56.
11. Seeram E, Seeram D. Image postprocessing in digital radiology: a primer for technologists. J Med Imaging Radiat Sci. 2008;29(1):23–43.
12. Siegel E, Reiner B. Radiology reading room design: the next generation. Appl Radiol. 2002:11–6.
13. DICOM: Digital imaging and communications in medicine. https://www.dicomstandard.org. Accessed Jan 2018.
14. Integrating the Healthcare Enterprise (IHE). IHE IT Infrastructure (ITI). Technical framework. Volume 1. IHE International; 2017.
15. IHE: Integrating the Healthcare Enterprise. https://www.ihe.net/About_IHE/. Accessed Jan 2018.
16. Roth CJ, Lannum LM, Dennison DK, Towbin AJ. The current state and path forward for enterprise image viewing: HIMSS-SIIM collaborative white paper. J Digit Imaging. 2016;29:567–73.
17. HIPPA for Professionals. US Department of Health and Human Services. https://www.hhs.gov/hipaa/for-professionals/index.html. Accessed Jan 2018.
18. HIMMS Definition of Interoperability. HIMSS Dictionary of healthcare information technology terms, acronyms and organizations. HIMSS Electronic Health Record Association, 2010; Appendix B, p190, original source.
19. Roth CJ, Lannum LM, Persons KR. A foundation for enterprise imaging: HIMSS-SIIM collaborative white paper. J Digit Imaging. 2016;29:530–8.
20. Roth CJ, Lannum LM, Joseph CL. Enterprise imaging governance: HIMSS-SIIM collaborative white paper. J Digit Imaging. 2016;29:539–46.
21. Towbin AJ, Roth CJ, Bronkalla M, Cram D. Workflow challenges of enterprise imaging: HIMSS-SIIM collaborative white paper. J Digit Imaging. 2016;29:574–82.
22. Webopedia. Cloud computing. https://www.webopedia.com/TERM/C/cloud_computing.html#delivery. Accessed Feb 2018.

Medical Imaging Informatics: An Overview

10

Abstract

Imaging informatics (II) has replaced the term medical imaging informatics (MI) and is now commonplace in the imaging community. The Society of Imaging Informatics in Medicine (SIIM) states that the "science of imaging informatics is the study and application of processes of information and communications technology for the acquisition, manipulation, analysis, and distribution of image data." This chapter describes the essential technologies such as the fundamentals of computers and communication technologies that are the building blocks for an understanding of II. In addition, relevant elements of the picture archiving and communication systems (PACS), the radiology information system (RIS), and the electronic health record (EHR) are highlighted in terms of definitions and components. Furthermore, an overview of system integration and information technology security is provided. Finally, emerging topics of II such as cloud computing, Big Data, artificial intelligence, machine learning, and deep learning are briefly described. Since the latter four topics, Big Data, artificial intelligence, machine learning, and deep learning, are in their infancy in terms of development and implementation, major definitions of each have been quoted from the experts, so as not to detract from the original meaning.

Big Data is characterized by four Vs: volume, variety, velocity, and veracity. While volume refers to the very large amount of data, variety deals with a wide array of data from multiple sources. Furthermore, velocity addresses the very high speeds at which the data is generated. Finally, veracity describes the uncertainty of the data such as the authenticity and credibility. The definitions for AI, machine learning, and deep learning are as follows (see appropriate citations in the text):

- AI is the "effort to automate intellectual tasks normally performed by humans."
- Machine learning is "a set of methods that automatically detect patterns in data, and then utilize the uncovered patterns to predict future data or enable decision making under uncertain conditions."
- Deep learning algorithms are "characterized by the use of neural networks with many layers."

As these emerging topics in imaging informatics evolve, they will gain acceptance and become useful tools in medical imaging technologies.

10.1 Introduction

The term "informatics" has been derived from the term "information." For computer applications in general, information refers to useful, meaningful,

E. Seeram, *Digital Radiography*, https://doi.org/10.1007/978-981-13-3244-9_10

and organized data. In other words, computers convert data into information, to be used by humans for solving problems in the real world.

The computer applications in medical imaging described in this book include modalities such as computed radiography (CR), digital radiography (DR) using flat-panel digital detectors, digital fluoroscopy (DF), digital mammography (DM), and digital tomosynthesis (DT), which have been described in detail in Chaps. 3, 4, 6, 7, and 8, respectively. These digital imaging technologies acquire raw data from the patient that are processed by computers to produce useful information. This information is of course the images that radiologists use to make a diagnosis of the patients' medical condition. Furthermore, the topic of digital image processing (Chap. 2) is a mandatory component of informatics. In addition, the computer processes textual and image data to produce useful information that is used to interpret images in an effort to manage the patient's medical condition. Subsequently, this information is sent to the picture archiving and communication system (PACS), or the vendor neutral archive (VNA), or to the enterprise imaging (EI) system, for display, viewing, and image processing, storage, and communication to remote sites to other related individuals who play an integral role in the medical management of the patient. Combining these technologies into one useful system leads to what is now referred to as medical imaging informatics (MII). Within the imaging community, a formal definition of MII has been offered by the Society of Imaging Informatics in Medicine (SIIM). SIIM states that the "science of imaging informatics is the study and application of processes of information and communications technology for the acquisition, manipulation, analysis, and distribution of image data" [1]. MII has received increasing attention in the literature and in clinical practice that, in their recent Strategic Plan 2017–2020, SIIM's vision and mission are to improve patient care through II and to support the vision through education research and innovation, respectively [2].

The purpose of this chapter is to outline the major elements of II, the application of information technology to imaging. In particular, this chapter will illustrate the contributions of various technologies that enable PACS and digital imaging to the evolution and growth of a new medical subspecialty for the twenty-first century popularly referred to as imaging informatics (II). Emerging topics of II such as Big Data, cloud computing, artificial intelligence, machine learning, and deep learning will be briefly outlined.

10.2 Information Technology

The production of vast amounts of information in all "walks of life," such as the National Aeronautics and Space Administration (NASA) space program, Internet vocations, and healthcare to medical imaging, has created a need for effective and efficient use and management of this information. This task is accomplished through the use of information technology or IT as it is popularly referred to.

10.2.1 Definition

IT is currently commonplace in medicine and is routinely used in hospitals, in almost all departments. IT now plays an important and significant role in the digital imaging department. Digital devices must be able to communicate with each other regarding images and textual information, and this function requires the use of IT.

There are several definitions of IT, but one that is perhaps best suited to this chapter is one offered by TechTarget [3]. They define IT as "the use of any computers, storage, networking and other physical devices, infrastructure and processes to create, process, store, secure and exchange all forms of electronic data" [3]. While computer technology deals with the structure and function of a computer and how it can be used to solve problems, computer communication technology, on the other hand, deals with the use of electromagnetic devices and systems for communicating over long distances [4].

It is not within the scope of this book to describe the details of computer and computer

communication technologies; however, a brief review of the essential elements of each technology is worthwhile for technologists. For further details of these two technologies, the reader should refer to books dedicated to computer and science [5–7].

10.2.2 Computer Technology Basics

A computer is an electronic machine for solving problems, through the use of a number of characteristic elements. A computer system, on the other hand, consists of at least three elements: hardware, software, and computer users [8, 9]. While hardware refers to the physical components of the machine, software deals with the instructions that make the hardware work to solve problems. People are essential to computer systems because they design, develop, and operate hardware and software.

The organization of a computer system is shown in Fig. 10.1. The following points are noteworthy.

1. Computers consist of various electronic devices for converting data (not useful) to information (useful) that can be used to solve problems.
2. A computer is made up of hardware and software. While the hardware refers to the physical components used to process data, software refers to computer programs (algorithms) that direct all hardware components to solve problems and thus derive useful practical solutions. Hardware devices include the keyboard; mouse; central processing unit (CPU); memory chips such as random access memory (RAM); storage devices, such as magnetic disks and tapes and optical disks; and monitors and printers. Software, on the other hand, refers to computer programs (algorithms) and includes system software and application software. While system software refers to operating systems such as Windows 10, macOS, iOS (mobile operating system for Apple Inc.), UNIX, and LINUX, application software refers to algorithms that perform specific tasks. These include, for example, Microsoft Word® for word processing and Adobe Photoshop® for image processing and manipulation.
3. Computers use at least five fundamental operations (Fig. 10.1) to process data and

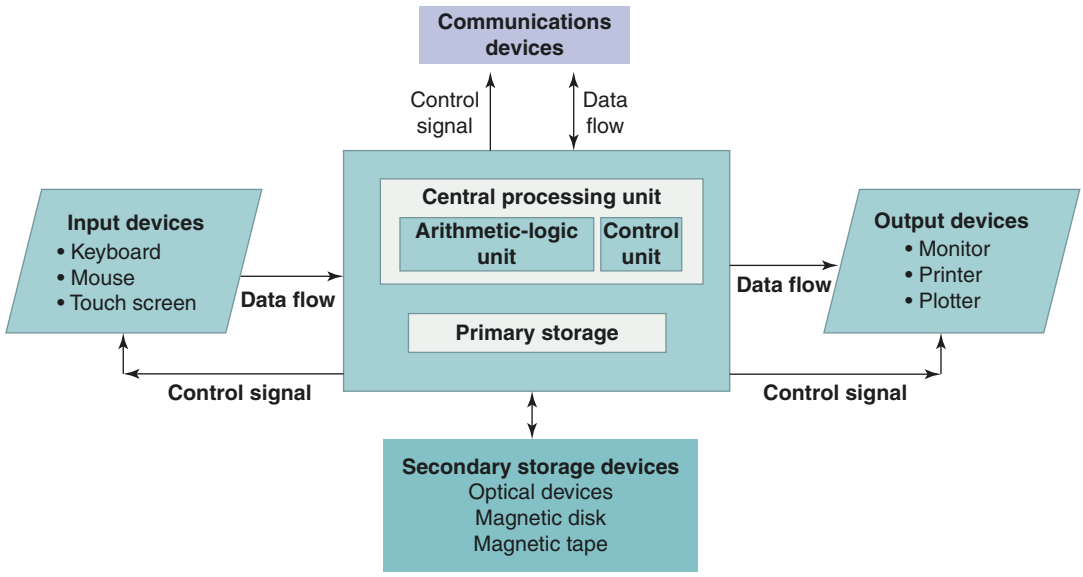

Fig. 10.1 The functional connection of major components of a computer system consists of input and output devices, central processing unit, secondary storage, and communication devices. See text for further explanation

disseminate information. These include in a linear order:

(a) Input. This refers to data entry using hardware devices such as a keyboard or a mouse, etc.
(b) Processing. This involves the use of hardware devices such as the CPU to process data entered into the computer. Processing converts data to information that can be used by human observers to solve problems, such as diagnosing a patient's medical condition.
(c) Storage. Once information is obtained, it can be stored temporarily inside the computer (internal storage or memory), or it can be stored permanently outside the computer (external storage). External storage can hold a larger amount of information than internal storage.
(d) Output. The results of computer processing can be displayed on monitors, for viewing by a human observer. For example, computed radiography data from the imaging plate is processed and displayed on a monitor as an image, for viewing and interpretation by a radiologist. Output images can be post-processed to enhance the interpretation skills of the radiologist.
(e) Communications. Computer data and output information (e.g., images) can be transmitted to other individuals anywhere in the world, with the use of the Internet, for example, as the communication vehicle.

10.2.3 Communication Technology Basics

Communication technology dates back to 1592 with the invention of the first newspaper in Italy. Popular devices that make use of communication technology include telephones, radios, movie cameras, televisions, cell phones, WebTV, Internet telephones, wireless services such as Wi-Fi (wireless fidelity), and Bluetooth, for example.

Computer communication technology consists of the following key elements:

- Modem. This is a contraction for *mo*dulate/*dem*odulate. The modem is used to send and receive signals from computers, by converting a digital signal from the computer (modulate) into an analog signal. This analog signal is transmitted over a communication link (e.g., phone line) to receiving computer. The analog signal must be converted back to a digital signal (demodulate) for processing by a digital computer. Modems can be external or internal to the computer and can transmit data in bits per second (bps). The bps will vary depending on the type of technology used.
- Communications media. Data transmission requires some sort of medium to do the task. The phone line is one such medium. Other communications media include wire pairs (twisted pairs), coaxial cables, fiber optics, microwave, and satellite transmission. Wireless transmission is also used to send data using radio frequencies to connect various physical devices. Protocols are essential tools for devices to communicate with each other. Protocols are rules for the exchange of data between the computer itself and a terminal or between two computers and are addressed by the hardware and software [8, 9]. The protocol used with Internet communications is the Transmission Control Protocol/Internet Protocol (TCP/IP). Standards also help to facilitate communication among different systems. DICOM (Digital Imaging and Communications in Medicine) is an example of one standard used in medical imaging to address in particular communication of images.
- Communication networks. A network allows several devices (e.g., computers) to be connected together in an effort to utilize data and information. The layout or configuration of the network is called a topology. Three typical networks, the bus, star, and ring networks, were described in Chap. 9 on PACS. A network consists of several components including a host computer and nodes (any device coupled to the network), packets (block of data to be transmitted), protocols, and other devices such as hubs, switches, bridges,

routers, and gateways. There are several types of networks based on the size of the network ranging from the smallest, the local area network (LAN), to the largest, the wide area network (WAN). The metropolitan area network (MAN) falls in between the LAN and the WAN. Apart from wired communications media described above, wireless communications media have become commonplace. These media include infrared transmission broadcast radio, microwave radio, and communications satellite. These media utilize different frequencies from the radio-frequency spectrum of the electromagnetic spectrum.

- Computer communication technology also involves topics such as cyber-threats, security, and safeguard issues; however, these topics are not within the scope of this book.

Digital medical imaging utilizes several aspects of IT, and therefore it is important for the technologist to have a general understanding of the essential concepts of IT. IT now has a great impact on the practice of medicine and more importantly on digital imaging. For example, IT is applied to information systems, standards for communicating both textual data and image data sets, communication networks, web technology, and privacy, confidentiality, and security of medical information, to mention only a few.

10.3 What Is Informatics?

As noted above, the term "informatics" is closely linked with IT. The term dates back several decades, and its origin can be traced back to Russian writings [10]. Informatics refers to the process of changing data to information, and therefore computers and communication technologies are central elements to the process.

10.3.1 Informatics Subspecialties

Informatics is now applied to several disciplines and has led to new subspecialties such as medical informatics, healthcare informatics, biomedical informatics, nursing informatics, and imaging informatics (radiology informatics), cardiology informatics, clinical informatics, and so on.

10.3.2 Healthcare Informatics/ Medical Informatics

The application of information technology to healthcare or medicine is referred to as healthcare informatics and medical informatics, respectively. There are several definitions in the literature and on various medical informatics websites; however, Healthcare Information and Management Systems Society (HIMSS) has defined both terms as "the interdisciplinary study of the design, development, adoption, and application of IT-based innovations in healthcare services delivery, management, and planning" [11].

10.3.3 Scope of Health Informatics

The scope of health informatics covers a wide and varied range of topics. Several textbooks [12–15], on health informatics, for example, provide some direction in this regard. In general, the common themes that emerge are centered on major components: core concepts (basic concepts of informatics relating to models, data, information, and system theory), information science and technology, computer science, ethical and legal aspects of health informatics, healthcare information systems, the electronic health record, health data analytics, security issues, data mining and machine learning, artificial intelligence, decision support, Internet technologies, and so forth.

Health information systems is an important topic and includes a study of clinical information systems (such as nursing information systems, laboratory information systems, cardiology information systems, and radiology information systems) and administrative systems which deal with such functions as patient registration and scheduling. Additionally, topics relating to information systems include training, security and confidentiality issues, system integration, the

electronic health record, regulatory issues, and disaster recovery, to mention only a few.

Finally, the topic of specialized applications in a healthcare informatics curriculum will deal with the use of computers in education and research. In addition, telehealth concepts and issues, as well as decision support systems, are mandatory topics in healthcare informatics.

10.4 Imaging Informatics

Medical imaging includes several modalities such as diagnostic radiography fluoroscopy, computed tomography, magnetic resonance imaging, medical sonography, and nuclear medicine. Furthermore, radiation therapy and nuclear medicine incorporate the CT imaging into its practice. In radiation therapy the CT scanner is now being used as an integral part of CT simulation to accurately define and localize tumors in a patient to aid radiation therapy planning and delivery. In nuclear medicine, CT scanners are now a part of the equipment to create hybrid images. A major objective of this approach is to combine the anatomical details of CT with the functional role of nuclear medicine images. It is clear that medical imaging plays a significant role in healthcare since it is used for diagnosis, assessment and planning, guidance of procedures, communication, education, training, and research.

10.4.1 Definition

The application of IT to medical imaging is referred to as radiology informatics or medical imaging informatics (MII) or imaging informatics (II). The latter term is now commonplace and will be used throughout this chapter. Imaging informatics is a new subdiscipline in radiology, and it was proposed several years ago by Kulikowski [16], who felt that the tasks performed by computers in a digital imaging department form the basis for this subspecialty. In particular, these tasks include digital image acquisition, digital image processing and image display, image storage and archiving, computer networking, and image transmission. Later the use of picture archiving and communication systems (PACS) has become commonplace in digital imaging all over the world and is continuing to evolve. The use of PACS is now a routine part of the duties of the technologist, and its overall framework was described in detail in Chap. 9.

Imaging informatics has been evolving through the years, as digital imaging departments become more and more efficient and commonplace. For example, several authors have described evolving nature of imaging informatics and its role in the twenty-first century [17, 18].

10.4.2 Framework for II

A framework for a PACS-based multimedia II has been conceptualized and described by Huang [19], a notable expert in the area of PACS and imaging informatics. Dr. HK Huang FRCR (Hon.), FAIMBE, is a professor of Radiology and Biomedical Engineering; director, Division of Imaging Informatics, Department of Radiology; director, MS Program, Medical Imaging and Imaging Informatics, Department of Biomedical Engineering, University of Southern California, Los Angeles; chair professor of Medical Informatics, the Hong Kong Polytechnic University; and honorary professor, Shanghai Institute of Technical Physics, the Chinese Academy of Sciences. In his classic textbook, titled *PACS-Based Multimedia Imaging Informatics*, Dr. Huang describes basic principles and applications in four parts. While Parts 1 and 2 deal with "the Beginning–Retrospective and Medical Imaging, Industrial Guidelines, Standards, and Compliance," Parts 3 and 4 address "Informatics, Data Grid, Workstation, Radiation Therapy, Simulators, Molecular Imaging, Archive Server and Cloud Computing and Multimedia Imaging Informatics, Computer-Aided Diagnosis (CAD), Image-Guide Decision Support, Proton Therapy, Minimally Invasive Multimedia Image-Assisted Surgery, Big Data". Interested readers should refer to this book for details of these topics. These topics and their organization provide the basis for knowledge

content that is essential to an imaging informatics curriculum, not only for radiologists and technologists but for medical physicists, biomedical engineers, and IT personnel. Such curriculum elements and associated credentialing will be highlighted later in the chapter.

10.5 PACS Technology

As mentioned earlier, IT involves two technologies: computer technology and communication technology. PACS is an example of an electronic system that is based on the use of computer and communication technologies. PACS therefore is an informatics-rich environment [18]. The essential elements of PACS were described in Chap. 9. In review, the major components of PACS include digital image acquisition modalities, a computer network database server, an archival system, and a soft-copy display workstation. In addition, a PACS is often connected to health information systems such as a radiology information system (RIS) and a hospital information system (HIS).

The images produced by the image acquisition modalities (CR, DR, CT, MRI, and so on) are sent to the PACS for storage and archiving and subsequent transmission to remote locations. In this regard, a computer network is used for image transfer using the DICOM standard. The images are stored in the database server (image server), the "brain" of a centralized PACS. Images are stored in an archival system consisting of short-term, long-term, or duplicate storage. Storage devices range from magnetic disks, optical disks, magneto-optical disks, and digital videodisks.

Display systems in a PACS environment fall into two categories: cathode ray tubes (CRTs) and liquid crystal displays (LCDs). The goal of image display is to facilitate image interpretation by the radiologist. The display monitor is an essential component of the PACS workstation. A display workstation used by radiologists features various tools used to enhance productivity. These tools are all informatics-related tools, such as prefetch algorithms, hanging protocols, image processing algorithms, and decision support tools.

The final component of a PACS involves access and distribution of images in the PACS to interested parties, not only in the radiology department but to others throughout the healthcare enterprise. While the former is referred to as a traditional radiology-centric PACS model, the latter is called the enterprise-wide image distribution model. Within a PACS environment, it is mandatory that the PACS be integrated with information systems, since these systems contain data and information related to the patient whose images are stored in the PACS.

10.6 Health Information Systems

An information system is a computer-based system used to collect and process data to provide its users with information needed for problem-solving and decision-making. A healthcare information system and hospital information system (HIS) are systems used within a hospital or enterprise that support and enhance the care and management of patients.

There are two major component systems of the HIS: clinical information systems (CISs) and administrative information systems (AISs). CISs include a number of information systems such as order entry system, monitoring system, nursing information system, laboratory information system, and a radiology information system (RIS). AISs include information systems dealing with registration, scheduling, financial payroll and human resources, quality assurance, and contract management. It is not within the scope of this book to describe these information systems; however, brief elements of the RIS will be highlighted to illustrate that the RIS is an essential component of imaging informatics.

A few characteristic and essential features of an RIS are as follows:

- Order entry: requested radiology examinations ordered by physicians are entered into the system. This is also known as registration of patients.
- Scheduling of various imaging procedures.

- Report generation: radiologists enter image interpretation reports. These reports are obviously linked to the appropriate patient and examination.
- Billing preparation: costs associated with various procedures done on patients are generated and sent to a billing system.
- Other related functions: these include functions such as quality assurance, inventory monitoring, and statistical analysis as well as communications that are important to the daily operations of the imaging department.

It is clear that the information contained in the RIS can be shared with the HIS, so that physicians in the hospital, for example, can access related information on their patients from their offices if required.

10.7 The Electronic Health Record: Brief Overview

With the widespread use of HIS, RIS, and PACS, the paper-based medical record is being replaced by the electronic health record (EHR) to overcome the limitations of paper-based records. One such major limitation is that the paper-based record cannot include images (such as those stored in the PACS). Furthermore, the paper-based record lacks the mechanism to make practical use of decision support and other related systems

10.7.1 Definition and Components

HIMSS states that the EHR "is a longitudinal electronic record of patient health information generated by one or more encounters in any care delivery setting. Included in this information are patient demographics, progress notes, problems, medications, vital signs, past medical history, immunizations, laboratory data and radiology reports. The EHR automates and streamlines the clinician's workflow. The EHR has the ability to generate a complete record of a clinical patient encounter—as well as supporting other care-related activities directly or indirectly via interface—including evidence-based decision support, quality management, and outcomes reporting" [20].

The EHR contains a vast amount of medical information, data about the patient (e.g., demographic information), and of course other related health information, such as images from medical imaging procedures. The EHR contains data and information such as order entry, clinical documentation, data repository, decision support, results reporting, and clinical messaging and e-mail. The EHR provides a wide range of benefits not only to healthcare professionals including physicians but also benefits that are available to the healthcare enterprise. Such benefits, for example, include improved data integrity and quality of care, as well as increased productivity and satisfaction for healthcare providers. The EHR will not be described further in this book; however, the interested student may refer to the textbook by Nelson and Staggers [12].

10.8 System Integration Overview

Integration is an important concept in imaging informatics, for it deals with the notion that to be effective and efficient, imaging modalities (data acquisition devices), PACS, and information systems must be able to communicate (exchange date) with each other, in a seamless manner.

10.8.1 Requirements for System Integration

There are several requirements for effective and efficient system integration. These include, and interface, data dictionary, uniform language, master patient index (MIP), and communication (data exchange) standards.

An interface serves to connect systems so that they can exchange data. The early interfaces were point-to-point interfaces because they allowed two systems to communicate with each other. Today, more sophisticated interfaces called interface engines that are computer programs

(software) facilitate the exchange of data among several systems. Such communication is made possible through the use of a clinical data repository and a technique referred to as mapping. The term "mapping" is used to refer to terms used in one system that must match comparable terms in another system. While a data dictionary provides the definition of terms use in the particular enterprise (e.g., healthcare enterprise), the MIP is a database that contains all information (e.g., demographics) about the patient and plays an important role in identifying and locating the records belonging to the patient under study.

The use of standards for data exchange in imaging informatics is a mandatory concept for radiologists and technologists, as well as related personnel working in the imaging department (e.g., physicists and biomedical engineers and IT personnel). As noted earlier in the chapter, the two standards that are used in the imaging department to communicate images and textual data are DICOM (Digital Imaging and Communications in Medicine) and HL-7 (Health Level-7), respectively. While DICOM has been developed specifically for communication of images from the image acquisition devices to the PACS and from the PACS to others within the enterprise, HL-7 has been developed to handle text data exchange among various information systems such as the RIS and HIS.

Finally, it is important to note that the Integrating the Healthcare Enterprise (IHE) initiative (Chap. 9) is also an important imaging informatics topic. This initiative dates back to 1998 through a joint venture between the Radiological Society of North America (RSNA) and the Healthcare Information and Management Systems Society (HIMSS). The overall objective of the IHE initiative is to ensure that image acquisition devices, PACS, and information systems from different vendors communicate (exchange data) with each other in a seamless fashion.

10.9 IT Security Fundamentals

As healthcare informatics continues to develop and become more and more commonplace, the topic of security becomes increasingly important. In this regard, for example, the Health Insurance Portability and Accountability Act (HIPAA) has drawn serious attention to the need for system security. In addition, HIPPA security rule requires that healthcare institutions determine threats and take action to protect information.

10.9.1 What Is Security?

The protection of information systems and maintenance of system integrity are functions of information security (IS) which "is designed to protect the confidentiality, integrity and availability of computer system data from those with malicious intentions. **C**onfidentiality, **I**ntegrity and **A**vailability are sometimes referred to as the CIA Triad of information security. This triad has evolved into what is commonly termed the Parkerian hexad, which includes confidentiality, possession (or control), integrity, authenticity, availability and utility" [21]. Security allows authorized users to have access to the information and prevents unauthorized users from accessing such information.

The topic of IT security is wide and varied and cannot be described here in any details. This section will introduce the student to the nature of computer security and to demonstrate that it is an important component of imaging informatics.

10.9.2 Security Threats

Threats to security fall into three categories, namely, social engineering attacks, hardware attacks, and software attacks. While social engineering attacks seek to obtain data and information from authorized users in the form of deception, hardware attacks refer to attacks on hardware components, such as the theft of a computer. Software attacks deal with attacks on operating systems and application software, for example. Few of these include viruses, worms, Trojan horses, malicious programs, and denial of service. The more ambitious student should refer to the works of Nelson and Staggers [12], Coiera [13], Peck [15], Ong [22], and Bhatia [23].

10.9.3 Security Methods

The vast array of security risks provides a significant rationale for security methods. Home computer users, for example, ensure that their computer system is protected to a certain degree through the use of antivirus and spyware detection programs. Similarly, organizations ensure that their IT systems are secured.

Security methods range from physical security, authentication, passwords, firewalls, and antivirus software to spyware detection software, encryption and keys, and wireless security measures. These topics are not within the scope of this book. For those who are interested in reading about the general basis for network security for the imaging enterprise, refer to the work of Huang [19].

10.10 The Benefits of Imaging Informatics

Imaging informatics evolved as the technologies of digital image acquisition, image processing, image display, storage, and communications became widespread and commonplace. The evolution of the EHR coupled with the information that accompanies a patient in radiology known as the entire patient entity (EPE) has led to the reality that imaging informatics provides benefits that are significant not only to the imaging department but to the healthcare enterprise.

The more educated individuals who work with patients for the purpose of restoring their health become, the more they are able to contribute efficiently, effectively, and reliably to patient care. Individuals trained in imaging informatics as a subspecialty of radiology provide such contribution, and in the end, the patient will benefit from their wisdom.

10.11 Certification in Imaging Informatics

The idea of going beyond and the PACS administrator certification have led to the creation of the Certified Imaging Informatics Professional (CIIP) located at http://siim.org/. The CIIP certification is a venture between the Society for Imaging Informatics in Medicine (SIIM), the American Board of Imaging Informatics (ABII), and the American Registry of Radiologic Technologist Technologists (ARRT). Details such as the mission, goals, eligibility criteria, examination development process, and content are available on the SIIM website. In 2007, the ABII was founded by the SIIM and the ASRT. The mission of ABII is "to enhance patient care, professionalism, and competence in imaging informatics. ABII has created and manages the Imaging Informatics Professional (IIP) certification program and awards the Certified Imaging Informatics Professional (CIIP) designation to qualified candidates" [24].

10.12 Emerging Concepts in Imaging Informatics: An Overview

The medical imaging community has experienced an explosion of emerging computing topics at the Radiological Society of North America (RSNA) meetings in Chicago held in 2017 and 2018. Few of these emerging topics which will have an impact on the practice of medical imaging include cloud computing, Big Data, artificial intelligence (AI), machine learning, and deep learning. In this section, an overall brief overview of each will be presented, including formal authoritative definitions which will be quoted so as not to detract from the actual meaning of these terms.

10.12.1 Cloud Computing

The above description of VNAs, enterprise imaging systems, and the EHR and their limitations with respect to data archiving and management (movement and processing of large imaging data sets) coupled with the increasing growth of imaging data and efforts to collaborate across the

healthcare enterprise provide a rationale for looking at other approaches for efficient and effective data interchange. One such approach or model is cloud computing.

What is cloud computing? Several definitions exist in the literature; however the one that will be highlighted in this chapter is from the National Institute of Standards and Technology (NIST), an organization founded in 1901 and is now a component of the US Department of Commerce. NIST plays an active role in advancing measurement science, standards, and technology. NIST states that cloud computing is "a model for enabling ubiquitous, convenient, on-demand network access to a shared poll of configurable computing resources that can be rapidly provisioned and released with minimal management effort or service provider interaction" [25]. In other words, it "refers to the access of computing resources through the Internet for purposes of data storage, aggregation, synthesis, and retrieval, together with the capacity to act on the data with computational algorithms and software packages" [26].

An important characteristic of cloud computing is that the Internet can be used to deliver a number of services to interested users. For medical imaging departments using PACS, VNAs, and enterprising imaging, this cloud computing stores the data from these local systems and allows such data to be accessed by any user, from anywhere in the world using the Internet. The present use and access to Gmail, Google Docs, and Dropbox is a good example of cloud computing in everyday activities that most individuals enjoy today.

The delivery of these services obviously requires hardware and software systems at the data centers which have been referred to as a "cloud" [27]. A cloud can be a private cloud, a public cloud, or hybrid cloud (both private and public) and offered through a number of cloud service models or cloud types. These include Software as a Service (SaaS), Platform as a Service (PaaS), and Infrastructure as a Service (IaaS) as illustrated in Fig. 10.2. Examples of SaaS, PaaS, and IaaS include Salesforce.com, Microsoft Office 365, Google G Suite, Dropbox, and Adobe Creative Cloud; Amazon Web Services, Microsoft Azure, IBM, and Google Cloud Platform; and Amazon Web Services, Microsoft Azure, IBM, and Google Cloud Platform, respectively. For more details on the meaning of common cloud service models, the interested reader should refer to reference [28].

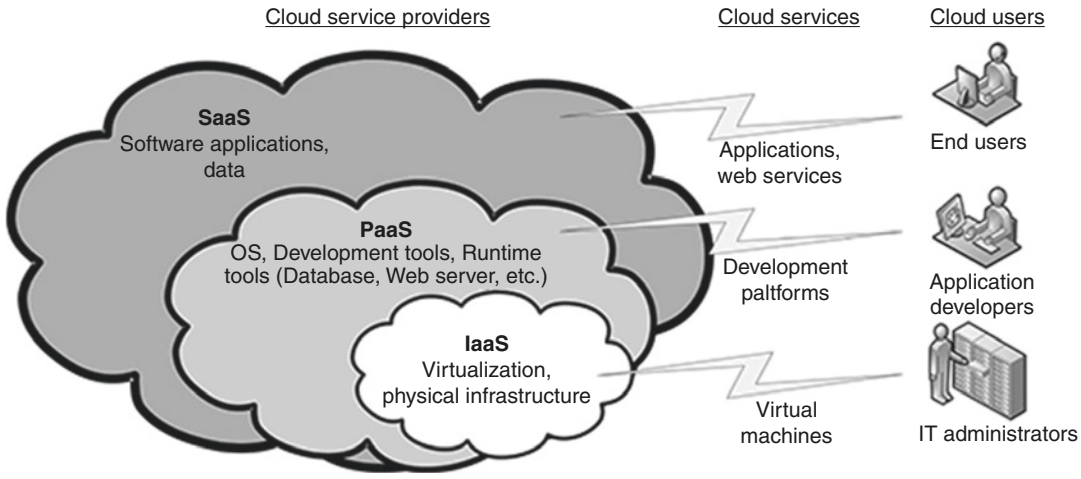

Fig. 10.2 Three types of cloud computing services (From Kagadis GC, Kloukinas C, Moore K, Philbin J, Papadimitroulas P, Alexakos C, Nagy PG, Visvikis D, and Hendee WR. Cloud Computing in Medical Imaging. *Medical Physics*; 2013, 40 (7). Reproduced by permission of John Wiley and Sons, UK). See text for further explanation

10.12.1.1 Cloud Computing and PACS

Cloud computing is gaining widespread attention in medical imaging and in particular PACS. The motivation for such interest rests upon not only the vast amount of data sets now available in medical imaging and the need for improved data management, image reconstruction, and image processing but for sharing and data storage [26]. Subsequently this will have a significant impact on PACS leading to what has been popularly referred to as "cloud-based PACS" or simply "Cloud PACS" which "is a PACS system that is located in the cloud and is accessible to both users and administrators though Internet-based user interfaces. Cloud PACS offer the promise of both location and device independence. That is, a Cloud PACS can be accessed by the user from any location, with the assumption that it has sufficient network connectivity and an appropriate connected device that has sufficient display characteristics" [26].

There are in general three components of Cloud PACS, that is, cloud-based image visualization, cloud-based workflow, and cloud-based image archive. Furthermore, ethics and security issues are extremely significant in the cloud computing environment. These issues are discussed, for example, by Kagadis et al. [26] who state that "the emerging technologies of cloud computing have already attracted several researchers, clinical administrators, and software developers to move medical image archives such as PACS onto the cloud, in order to improve manageability, accessibility, and storage availability" [26].

10.12.2 Big Data

There has been an explosion of digital data generated each day from several notable agencies, such as the National Aeronautics and Space Administration (NASA) program, social media (Facebook, Twitter), photos from cell phones for people around the world, Internet search engines such as Google, governments, healthcare, and last but not least medical imaging and radiation treatment planning technologies. These data are now popularly being referred to as "Big Data" which is also being referred to as an "emerging science" in imaging informatics. As noted by Kansagra et al. [29], "Big Data will transform the practice of medicine. Among different specialties, radiology-which has a mature IT infrastructure and many years of available digital data-is particularly well positioned to lead and benefit from these advances.....Among other applications, Big Data in radiology has the power to enable personalized image interpretation, discovery of new imaging markers, value quantification, and workflow characterization" [29].

10.12.2.1 Definitions

What is Big Data? Several definitions exist [30] from typical authoritative sources such as the following:

1. "Big Data consists of extensive data sets-primarily in the characteristics of volume, variety, velocity, and/or variability-that require a scalable architecture for efficient storage, manipulation, and analysis" [31].
2. Big Data: "an accumulation of data that is too large and complex for processing by traditional database management tools" [32].
3. "Big Data refers to extremely complex data sets characterized by the four Vs: *Volume*, which refers to the sheer number of data elements within these extremely large data sets; *Variety*, which describes the aggregation of data from multiple sources; *Velocity,* which refers to the high speed at which data is generated; and *Veracity*, which describes the inherent uncertainty in some data elements" [29].
4. A recent general agreement among experts is that Big Data "represents the Information assets characterized by such a High Volume, Velocity, and Variety to require specific Technology and Analytical Methods for its transformation into Value" [33].

10.12.2.2 Characteristics

Big Data is characterized by four Vs: volume, variety, velocity, and veracity as shown in Fig. 10.3. While volume refers to the very large amount of data, variety deals with a wide array of data from multiple sources. Furthermore, velocity

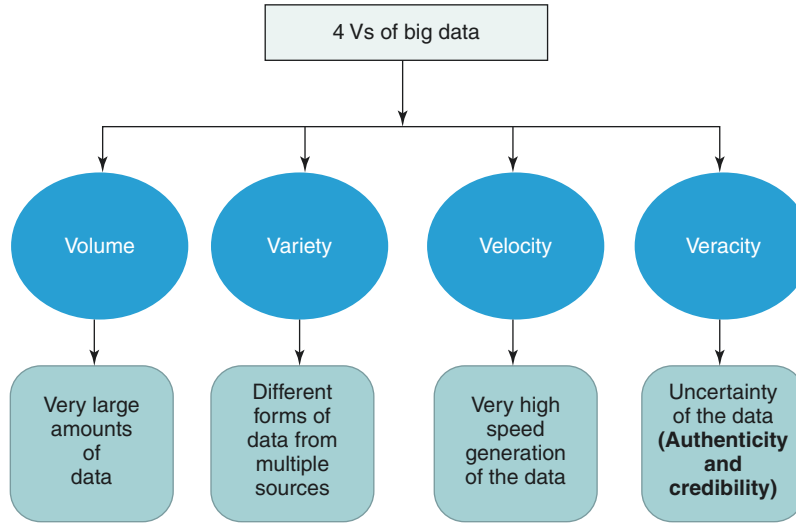

Fig. 10.3 Big Data is characterized by the four Vs; volume, variety, velocity, and veracity. See text for further explanation

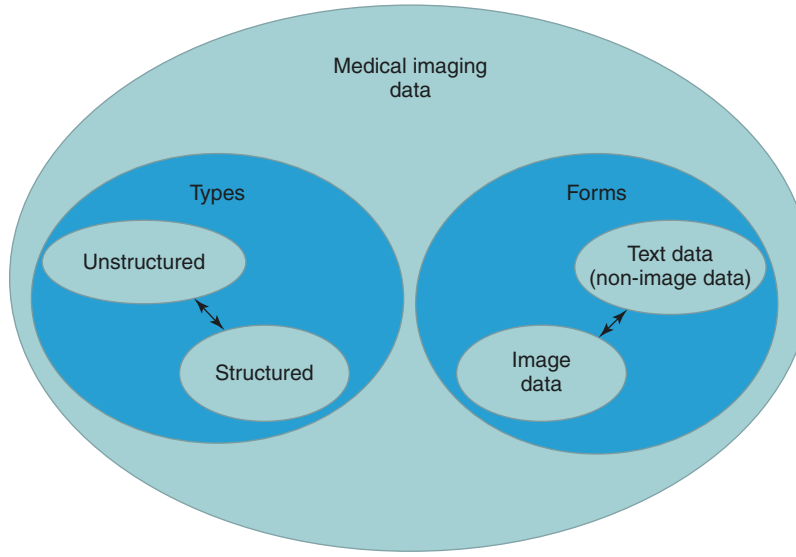

Fig. 10.4 Medical imaging data fall into two categories, namely, types and forms. While subsets of types of data include structured and unstructured data, subsets of forms of data include image data and non-image data

addresses the very high speeds at which the data is generated. Finally, veracity describes the uncertainty of the data such as the authenticity and credibility [29–31, 33, 34]. Medical imaging data can be categorized as types and forms as illustrated in Fig. 10.4. While types refer to unstructured and structured data, forms refer to image data and non-image (text) data [34]. Specifically, with respect to the four Vs of Big Data, the volume of data in imaging refers to the vast amount of image and non-image data stored in the PACS, VNAs, and RIS systems and includes images from digital radiography (DR), digital fluoroscopy (DF), CT including both routine and specialized CT examinations such as cardiac CT, magnetic resonance imaging (MRI), digital mammography (DM) including digital breast tomosynthesis (DBT), digital subtraction angiography (DSA), and nuclear medicine (NM) and diagnostic medical sonography (DMS). Velocity is the speed at which images from CT, MRI, DR, DF, DM, DBT, and DSA are sent to the PACS and later stored in VNAs. While variety in imaging refers to the different image types

generated by the different imaging modalities mentioned above, veracity deals with the uncertainty (authenticity and credibility) of the data. This means that in order to get meaningful results (new diagnostic information) from the analysis of the data, images must be artifact-free and be of diagnostic quality [34]. Furthermore, structured data also play as an important contributor of meaningful results after data analysis.

10.12.2.3 Big Data Preparation and Analytics

The generation of new diagnostic results from Big Data requires at least two steps as shown in Fig. 10.5. These include data preparation and data analytics. It is not within the scope of this chapter to explain details of these two operations since they require significant understanding of several subject matter areas, an important one being that of data science. In broad terms, the National Academies of Sciences, Engineering, and Medicine defines data science as "a multidisciplinary field that deals with technologies, processes, and systems to extract knowledge and insight from data and supports reasoning and decision making under various sources of uncertainty" [35].

In order for Big Data to be analyzed, it is essential that the raw data stored in PACS or VNAs be subject to various data preparation tasks and methods. While an example of a task is to locate, acquire, and ingest the data, an example of a method of data preparation is natural language processing [35]. Natural language belongs to the domain of computer science and, in particular, artificial intelligence (AI). Once the data has appropriately been prepared, it is now ready for the operation of data analytics. To reiterate, this topic is highly complex and therefore is not within the scope of this book; however, a definition is noteworthy. Such a definition is available from Davenport and Harris [36] who state that it is "extensive use of data, statistical and quantitative analysis, explanatory and predictive models and fact-based management to drive decisions and actions." Furthermore, data analytics involves various descriptive, predictive, and prescriptive analytics and various models such as statistical models which are central to data analytics [35]. Examples of predictive analytics include text analysis, data mining (knowledge discovery), and machine learning (teaching computers without the need for straightforward programming), all terms which have now become commonplace in imaging informatics.

10.12.2.4 Big Data Technologies

In a literature review of Big Data applications in healthcare, Luo et al. [37] identified several technologies used together for Big Data analytics. These include data mining tools, cloud computing technologies, and AI. The authors also identified several parallel computing models (MapReduce™ by Google and Hadoop® by Apache) and specifically state that several Big Data applications use cloud computing technologies.

10.12.2.5 Big Data Applications

Big Data applications have been identified and discussed in the literature [29, 30, 34, 37, 38]. In broad terms, these applications range from bioinformatics, clinical informatics, imaging informatics, and public health and information [37]. Specific applications of Big Data in imaging informatics fall under several categories such as efficient treatment and diagnosis processes for the patient, population studies using Big Data

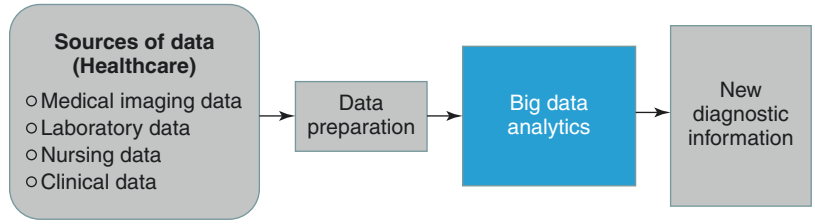

Fig. 10.5 Two steps needed to generate new diagnostic results from Big Data include data preparation and data analytics. See text for further explanation

and their effects, research and education applications of Big Data, departmental efficiency solutions and cost reduction, and safety improvements [38]. Furthermore, other specific applications such as scheduling of scans, creating patient-specific personalized scanning protocols, radiologist decision support, emergency reporting, and virtual quality assurance for the radiologist have been discussed by Kharat and Singhal [34]. Additionally, Kansagra and others [29] have discussed applications in image interpretation and management, identification of new diagnostic information in imaging data, and clinical outcomes, cost of care, assessment, and optimization of workflows. Finally, in terms of using analytics in imaging, Belle and others [39] have described the use of medical image analysis, physiological signal processing, and genomic data processing.

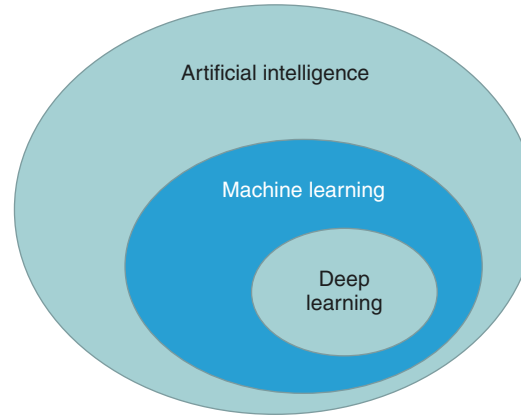

Fig. 10.6 AI is a branch of computer science and consists of two important subsets as shown in the Venn diagram. These subsets include machine learning and deep learning (From Chollet F. Deep Learning with Python. Shelter Island NY, Manning Publications Co. 2018. Courtesy of Manning Publications Co; Shelter Island NY)

10.12.3 Artificial Intelligence, Machine Learning, and Deep Learning

As mentioned above, several technologies used for Big Data analytics include data mining tools, cloud computing technologies, and artificial intelligence (AI). AI is a field of computer science, and its use is gaining attention in healthcare [40] and especially in medical imaging [41–44]. What is AI, what are its subsets, and what are their applications in medical imaging? AI is a branch of computer science and consists of two important subsets as shown in the Venn diagram in Fig. 10.6. These subsets include machine learning and deep learning, and as noted by Erickson et al. [45], "the use of machine learning in radiology has important implications for the practice of medicine, and it is important that we engage this area of research to ensure that the best care is afforded to patients. Understanding the properties of machine learning tools is critical to ensuring that they are applied in the safest and most effective manner" [45]. Already primers for radiologists on these topics have appeared in popular and credible radiology journals [40–44].

It is important to understand that the efforts involved in introducing and implementing these emerging technologies into the domain of medical imaging in particular mean that there are significant benefits to be realized. One such benefit is "to help radiologists provide a more accurate diagnosis, by delivering quantitative analysis of suspicious lesions, and may also enable a shorter time for reading due to automatic report generation and voice recognition, both of which are benefits that AI can provide in the clinical workflow" [44].

AI is an extensive subject and has its roots in philosophy, logic, mathematics, biology, cognitive science, neuroscience, and evolution. Furthermore AI consists of several areas such as expert systems, natural language processing, computer vision and scene recognition, neural computing, robotics and sensory systems, and intelligent computer-aided instruction. Therefore, a detailed description of AI is not within the scope of this book; however definitions will be quoted from the experts so as not to detract from the original meaning. In addition, applications in healthcare and medical imaging will be highlighted.

10.12.3.1 Definitions

The birth of AI can be traced back to 1950s when researchers (most notably two pioneers such as

Marvin Minsky of the Massachusetts Institute of Technology (MIT) and John McCarthy, at Stanford University) wanted to know if computers could think. Later several definitions of AI appeared in the literature [45–47].

One definition offered by Jiang et al. [40] is that AI "aims to mimic human cognitive functions." Another definition offered by Chartrand et al. [43] is that AI is "devoted to creating systems to perform tasks that ordinarily require human intelligence."

A most recent definition (2018) offered by Chollet [45] in his textbook on deep learning is that AI is the "effort to automate intellectual tasks normally performed by humans." As noted above, there are two categories of AI, machine learning and natural language processing. While the former operates on structured data (Fig. 10.4), the latter operates on unstructured data (Fig. 10.4) "to supplement and enrich structured medical data" [40]. Machine learning has been defined by Murphy [48] as "a set of methods that automatically detect patterns in data, and then utilize the uncovered patterns to predict future data or enable decision making under uncertain conditions." Additionally, Erickson et al. [41] point out that in machine learning, "algorithms are trained to perform tasks by learning patterns from data rather than by explicit programming." Furthermore in medical imaging, Erickson et al. [41] further explain that "if a machine learning algorithm is applied to a set of data (in our example, tumor images) and to some knowledge about these data (in our example, benign or malignant tumors), then the algorithm system can learn from the training data and apply what it has learned to make a prediction (in our example, whether a different image is depicting benign or malignant tumor tissue)."

The last term to be defined is that of deep learning. As mentioned earlier, deep learning is a subset of machine learning. Deep learning algorithms are "characterized by the use of neural networks with many layers" [43] and are "a special type of artificial neural network (ANN) that resembles the multilayered human cognition system" [44].

The conceptual difference between real neurons (derived from biology) and artificial neurons is illustrated in Fig. 10.7. For humans to think requires a complex network of neurons (units of the nervous system) using chemical and electrical signals. The neuron is made up of a cell body, dendrites, axon, and synapses (Fig. 10.7a). Incoming signals are transmitted from one neuron to the next via synapses. In Fig. 10.7b, the ANN is characterized by interconnected artificial neurons, where "each artificial neuron implements a simple classifier model that outputs a decision signal based on the weighted sum of evidences. Hundreds of these basic computing units are assembled together to establish the ANN. The weights of the network are trained by a learning

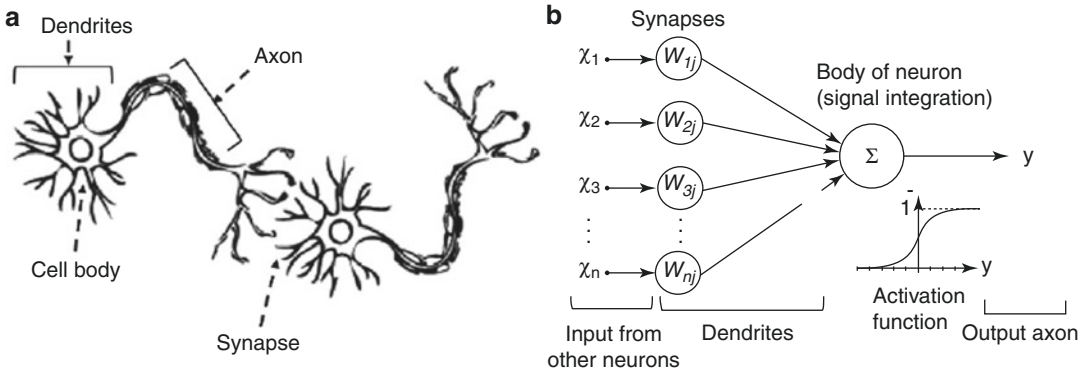

Fig. 10.7 The conceptual difference between real neurons (derived from biology) and artificial neurons (From Lee J-G, Jun S, Cho Y-W, Lee H, Kim GB, Seo JB, et al. Deep Learning in Medical Imaging: General Overview. Korean J Radiol 2017;18(4):570–584. Reproduced by permission)

algorithm, such as back propagation, where pairs of input signals and desired output decisions are presented, mimicking the condition where the brain relies on external sensory stimuli to learn to achieve specific tasks" [44]. Specific applications of deep learning in medical imaging will be listed in the next section. In everyday use, deep learning is being used in Siri® (Apple), Google Home, and Amazon's Echo. For more details of deep learning, the interested reader should refer to the papers by Chartrand et al. [43] and by Lee et al. [44].

10.12.3.2 Applications in Healthcare and Medical Imaging

Pannu [47] has identified and described the general areas of AI. These areas include language understanding, learning, and adaptive systems; problem-solving; perception; modeling; robotics; and games such as chess. Within the context of this chapter, AI has specific applications in healthcare and medical imaging.

In healthcare, Jiang et al. [40] surveyed the literature on the current status of AI, its future direction. They reviewed four areas of concern: the motivation for applying AI techniques to healthcare, the types of data subject to AI techniques, the AI techniques used, and the types of disease currently being examined in which AI techniques can be applied to generate useful and valid results. The data types include both structured and unstructured data in medical imaging, genetics, electrodiagnosis, monitoring, physiologic, disability evaluation, mass screening, and so forth. Popular AI techniques used are machine learning for structured data and natural language processing for unstructured data. Additionally, they identified major areas using AI tools such as cancer, neurology, and cardiology and reviewed ten disease types examined in the literature.

The applications of machine learning in everyday human activity include its use in banking, finance, marketing, and the Internet [42]. In medical imaging, a number of applications have been identified and discussed in the literature. For example, machine learning and deep learning applications in medical imaging range from image segmentation, image registration, automatic labeling and captioning, computer-aided detection and diagnosis to reading assistant and automatic dictation and integration with Healthcare Big Data: Towards Precision Imaging [41–44]. The integration with healthcare Big Data is linked to what has been referred to as precision medicine which "involves the prevention and treatment strategies that consider individual variability by assessing large sets of data, including patient information, medical imaging, and genomic sequences. The success of precision medicine is largely dependent on robust quantitative biomarkers. In general, deep learning can be used to explore and create quantitative biomarkers from medical big data obtained through internet of things, genetics and genomics, medical imaging, and mobile monitoring sources (Fig. 10.8). In particular, imaging is non-invasively and routinely performed for clinical practice, and can be used to compute quantitative imaging biomarkers. Many radiomic studies have correlated imaging biomarkers with the genomic expression or clinical outcome. Deep learning techniques can be used to generate more reliable imaging biomarkers for precision medicine" [44].

It is important to note that AI, machine learning, and deep learning are at the introductory stages of their development and applications in medical imaging. Furthermore, several experts [41–44, 49–51] have stated categorically that as these emerging topics in imaging informatics evolve, they will gain acceptance and become useful tools in medical imaging technologies. Imaging technology students and practicing technologists should make every effort to understand AI, machine learning, and deep learning and their applications in imaging informatics, in order to communicate effectively with radiologists and medical physicists and, later through more formal coursework on these topics, be able to assume leadership in imaging informatics. This chapter provides one small step in that direction. These tools can only help in restoring the health of patients.

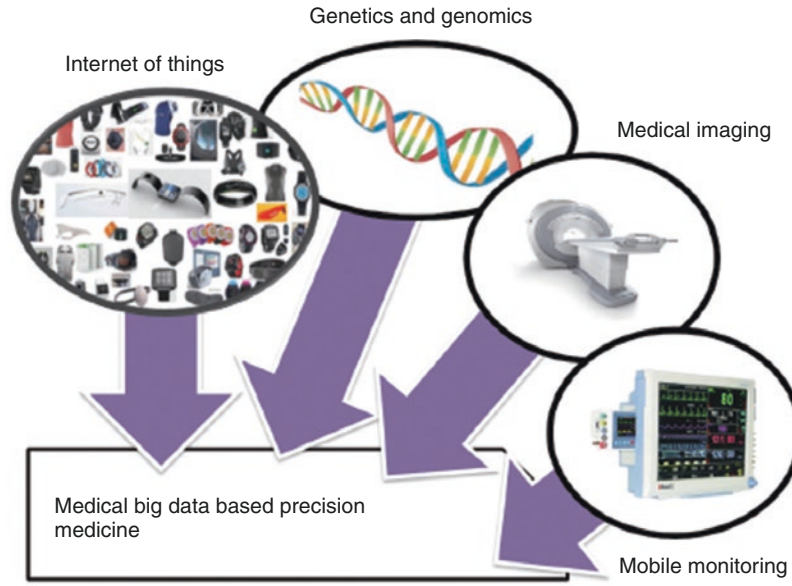

Fig. 10.8 The use of deep learning to explore and create quantitative biomarkers from medical Big Data obtained through Internet of things, genetics and genomics, medicinal imaging, and mobile monitoring sources (From Lee J-G, Jun S, Cho Y-W, Lee H, Kim GB, Seo JB, et al. Deep Learning in Medical Imaging: General Overview. Korean J Radiol 2017;18(4):570–584. Reproduced by permission)

References

1. Society of Imaging Informatics in Medicine (SIIM). What is imaging informatics? http://siim.org/ Accessed Feb 2018.
2. Society of Imaging in Medicine (SIIM). Innovating imaging informatics: strategic plan 2017-2020. SIIM. 2017:2–13.
3. Information Technology Definition. TechTarget: http://searchdatacenter.techtarget.com/definition/IT. Accessed Feb 2018.
4. Williams BK, Sawyer SC. Using IT: a practical introduction to computers and communications. New York: McGraw-Hill; 2012.
5. Subrata D. Computer science: a very short introduction. Oxford: Oxford University Press; 2016.
6. White C. Data communications and computer networks. 8th ed. Boston, MA: Cengage Learning; 2015.
7. Davis ER. Computer vision: principles, algorithms, applications, learning. 5th ed. London: Academic; 2017.
8. Hare K. Computer science principles: the foundational concepts of computer science. Doraville, GA; 2017.
9. Warford SJ. Computer systems. Burlington, MA: Jones and Bartlett Learning; 2016.
10. VanBemmel JH, Musen MA. Handbook of medical informatics. New York: Springer-Verlag; 1997.
11. HIMSS. What is health informatics? http://www.himss.org/health-informatics-defined. Accessed Feb 2018.
12. Nelson R, Staggers N. Health informatics: an interprofessional approach. 2nd ed. St Louis, MO: Elsevier; 2017.
13. Coiera E. Guide to health informatics. 3rd ed. Boca Raton, FL: CRC Press. p. 2015.
14. Mastrain K, McGonigle D. Informatics for health professionals. Burlington, MA: Jones Bartlett Learning; 2016.
15. Peck A. Clark's essential PACS, RIS and imaging informatics. Boca Raton, FL: CRC Press; 2018.
16. Kulikowski CA, Jaffe CC, editors. Focus on imaging informatics. J Am Med Informat Assoc. 1997;4(3):165–256.
17. Brandstetter B. Basics of imaging informatics. Radiology. 2007;243(3):656–67.
18. Brandstetter B. Basics of imaging informatics. Radiology. 2007;244(1):78–84.
19. Huang HK. PACS-based multimedia imaging informatics: basic principles and operations. Hoboken, NJ: Wiley; 2018.
20. HIMSS. What is the electronic health Record? http://www.himss.org/library/ehr. Accessed Feb 2018.
21. Technopedia. Information security definition. https://www.techopedia.com/definition/10282/information-security-is. Accessed Feb 2018.
22. Ong KR. Medical informatics: an executive primer. 3rd ed. Boca Raton, FL: CRC Press; 2015.
23. Bhatia D. Medical informatics. Delhi: PHI Learning Private Limited; 2015.
24. American Board of Radiology. Certification in imaging informatics. www.abii.org. Accessed Feb 2018.
25. Mell P, Grence T. The NIST definition of cloud computing. Special Publication. 2011;800-145.
26. Kagadis GC, Kloukinas C, Moore K, Philbin J, Papadimitroulas P, Alexakos C, Nagy PG, Visvikis D, Hendee WR. Cloud computing in medical imaging. Med Phys. 2013;40(7):070901-1–070901-10.

27. Armbrust M, Fox A, Griffith R, Joseph AD, Katz R, Konwinski A, Lee G, David Patterson D, et al. A view of cloud computing. Commun ACM. 2010;53:50–8.
28. Webopedia. Cloud computing. https://www.webopedia.com/TERM/C/cloud_computing.html#delivery. Accessed Feb 2018.
29. Kansagra AP, J-PJ Y, Chatterhee AR, Lenchik L, Chow DS, Prater AB, Yeh J, et al. Big data and the future of radiology informatics. Acad Radiol. 2016;23:30–42.
30. Morris MA, Saboury B, Burkett B, Gao J, Siegel EL. Reinventing radiology: big data and the future of medical imaging. J Thorac Imaging. 2018;33(1):4–16.
31. National Institute of Standards and technology (NIST). NIST big data interoperability framework: Volume 1, definitions—NIST.SP.1500-1.pdf. 2015. http://nvlpubs.nist.gov/nistpubs/SpecialPublications/NIST.SP.1500-1.pdf. Accessed 23 Feb 2018.
32. Merriam-Webster Dictionary. What is Big Data? https://www.merriam-webster.com/dictionary/big%20data. Accessed 21 Feb 2018.
33. De Mauro A, Greco M, Grimaldi M. A formal definition of Big Data based on its essential features. Libr Rev. 2016;65:122–35.
34. Kharat AT, Singhal S. A peek into the future of radiology using big data applications. Indian J Radiol Imaging. 2017;27(2):241–8.
35. National Academies of Sciences, Engineering, and Medicine. Strengthening data science methods for Department of Defence Personnel and Readiness Missions. Washington, DC: The National Academies Press; 2017.
36. Davenport TH, Harris JG. Competing on analytics: the new science of winning. Boston: Harvard Business Review Press; 2007.
37. Luo J, Wu M, Gopukumar D, Zhao Y. Big data application in biomedical research and health care: a literature review. Biomed Informat Insights. 2016;8:1–10.
38. Filonenko E, Seeram E. Big data: the next era of informatics and data science in medical imaging: a literature review. J Clin Exp Radiol. 2017;1:1–6.
39. Belle A, Thiagarajan R, Soroushmehr R, Navidi F, Beard DA, Najarian K. Big data analytics in healthcare. Biomed Res Int. 2015:1: 1–16.
40. Jiang F, Jiang Y, Zhi H, Dong Y, Li H, Ma S, et al. Artificial intelligence in healthcare: past, present and future. Stroke Vasc Neurol. 2017;2:230–43.
41. Erickson BJ, Korfiatis P, Akkus Z, Kline TL. Machine learning for medical imaging. Radiographics. 2017;37:505–15.
42. Wang S, Summers RM. Machine learning and radiology. Med Image Anal. 2012;16(5):933–51.
43. Chartrand G, Cheng PM, Vorontsov E, Drozdzal M, Turcotte S, Pal CJ, et al. Deep learning: a primer for radiologists. Radiographics. 2017;37:2113–213.
44. Lee J-G, Jun S, Cho Y-W, Lee H, Kim GB, Seo JB, et al. Deep learning in medical imaging: general overview. Korean J Radiol. 2017;18(4):570–84.
45. Chollet F. Deep learning with python. Shelter Island NY: Manning Publications; 2018.
46. Ertel W. Introduction to artificial intelligence. 2nd ed. Cham: Springer International Publishing AG; 2017.
47. Flasinski M. Introduction to artificial intelligence. Cham: Springer International Publishing; 2017.
48. Murphy KP. Machine learning: a probabilistic perspective. 1st ed. Cambridge: The MIT Press; 2012. p. 25.
49. Giger ML. Machine learning in medical imaging. J Am Coll Radiol. 2018;15(3):512–20.
50. Lakhani P, Prater AB, Hutson RK, Andriole KP, Dreyer KJ, Morey J, et al. Machine learning in radiology: applications beyond image interpretation. J Am Coll Radiol. 2018;15(2):350–9.
51. Pannu A. Artificial intelligence and its applications in different areas. Int J Eng Innovat Technol. 2015;10(4):79–84.

Continuous Quality Improvement for Digital Radiography

11

Abstract

Continuous quality improvement is a process to ensure that every employee plays a role in ensuring a quality product, and it includes two major elements, quality assurance (QA) and quality control (QC). While QA primarily deals with the administrative aspects of patient care and quality outcomes, QC addresses the more technical aspects of equipment performance. Both QA and QC play major roles in dose optimization that is a radiation protection principle to keep the dose as low as reasonably achievable (ALARA) without affecting the diagnostic quality of the image. QC is a common activity of technologists, medical physicists, and radiologists working in a medical imaging department, and it involves at least three essential tasks: that of acceptance testing, routine performance, and error correction. Furthermore acceptance criteria or tolerance limits for the recommended digital radiography QC tests, in which both qualitative and quantitative criteria may be used to assess image quality are provided.

A quality image is one that makes accurate diagnosis possible. This is known as "diagnostic quality." The quality of any image can be described in terms of spatial resolution, contrast resolution, noise, detective quantum efficiency (DQE), and artifacts. QA and QC activities demand an understanding of the processes and errors in digital radiography. A process map which is a flowchart of the steps involved in performing a DR exam can be created as a way to understand the interrelationships between activities within the imaging operation. There are at least 16 steps that begin with the arrival of the patient in the imaging department to the step (step 16) where the images are released to the picture archiving and communication system (PACS). The process map draws attention to errors in the association of demographic and exam information, errors that can be avoided by periodic testing, errors in performing the examination, and errors in the delivery of the images. Additionally, reject analysis is an essential element of QC activities, and it is a time-honored method for assessing and improving quality of imaging operations.

Examples of several QC tests that employ qualitative criteria for acceptance limits have been described briefly and include tests for dark noise, computed radiography imaging plate test for uniformity and spatial accuracy, and erasure thoroughness. Furthermore, ongoing QC is an essential element and process of any digital imaging department, and as such the American Association of Physicists in Medicine (AAPM) has provided guidance details of such activities such as reject image analysis, exposure analysis, and artifact identification. The AAPM, the American College of Radiology (ACR), and the Society for Imaging Informatics in Medicine (SIIM) identify and describe at least five individuals

E. Seeram, *Digital Radiography*, https://doi.org/10.1007/978-981-13-3244-9_11

who should play a significant role in QA/QA programs including the physician, a qualified medical physicist (QMP), registered radiologist assistant, radiologic technologist, and the imaging informatics professional.

Digital mammography is now an acceptable modality in most imaging departments, and the ACR provides *recommendations* for *full-field digital mammography quality control* in which they provide a list of 13 QC tests specifically for the technologist and 17 QC tests to be carried out specifically by the qualified medical physicist.

11.1 Introduction

There are several topics included in this chapter taken from Chap. 10 of the first edition of *Digital Radiography: An Introduction* published by Delmar Cengage Learning. That chapter was entitled "Quality Control for Digital Radiography" which was written by Charles Willis, PhD, DABR, Associate Professor, Department of Imaging Physics, The University of Texas MD Anderson Cancer Center, Houston, Texas. In this chapter, several topic areas have been included, in particular Sects. 11.4 and 11.7. Such materials are attributed to Dr. Charles Willis, PhD.

The maintenance of equipment is an integral part of the daily activities in radiology since it is considered a component of a *continuous quality improvement* (CQI) program. The notion of CQI was developed by the Joint Commission on Accreditation of Healthcare Organizations (JCAHO) in 1991. CQI ensures that every employee plays a role in ensuring a quality product. Prior to the introduction of the concept of CQI, other systems were in place to ensure quality patient care in hospitals. In medical imaging in particular, quality assurance (QA) and quality control (QC) programs are essential not only for optimizing the assessment and evaluation of patient care but also for monitoring the performance of equipment. Today QA and QC programs have become integral components in the daily operations of imaging departments and constitute an essential activity for radiologists, technologists, and medical physicists. Furthermore, various accrediting organizations, for example, the American College of Radiology (ACR); the Intersocietal Accreditation Commission (IAC), an organization dedicated to "improving healthcare through accreditation"™; and the Joint Commission (TJC), all require effective QC programs. For example, TJC requires a "ten-step monitoring and evaluation process" all intended to assure high quality of patient care and services [1]. Additionally, it is important to observe that academic imaging departments are also actively engaged in CQI programs using various tools and strategies [2].

11.2 Elements of CQI

11.2.1 Definitions of QA and QC

Quality assurance (QA) is a term used to describe systems and procedures for assuring quality patient care. It deals specifically with quality assessment, continuing education, the usefulness of quality control procedures, and the assessment of outcomes. QA deals with the administrative aspects of patient care and quality outcomes.

Quality control (QC), on the other hand, is a component of QA and refers specifically to the monitoring of important variables that affect image quality and radiation dose. QC deals with the technical aspects (rather than administrative aspects) of equipment performance. The purpose of the procedures and techniques of CQI, QA, and QC is threefold: to ensure optimum image quality for the purpose of enhancing diagnosis, to optimize the radiation dose to patients and reduce the dose to personnel, and to reduce costs to the institution.

11.2.2 Dose Optimization

Dose optimization is a radiation protection principle that has been described by the International Commission on Radiological Protection (ICRP), and it implies keeping the

dose as low as reasonably achievable (ALARA) without affecting the diagnostic quality of the image. All medical imaging departments operate within the ALARA philosophy. The ICRP defines at least two levels of optimization, namely, (a) the design and function of the X-ray imaging equipment and (b) imaging techniques and imaging protocols used during daily operation. While the former examines the design and function of the equipment to meet current radiation protection standards, the latter deals with the procedural and operational practices during the conduct of the examination [3].

The essential elements of a QA/QC program that play a role in dose optimization include responsibilities, QC principles and test procedures, and parameters for QC monitoring. In a typical QA/QC program, the radiologist in charge of the facility or a department assumes the primary responsibility for the entire program as identified by organizations such as The Joint Commission (TJC), Food and Drug Administration (FDA), and Radiation Protection Bureau-Health Canada (RPB-HC) [3].

Quality control involves a number of activities that are of significance. These activities range from acceptance testing to routine performance to error correction [1]. While *acceptance testing* is the first major step in a QC program and it ensures that the equipment meets the specifications set by the manufacturers, *routine performance* involves performing the actual QC test on the equipment with varying degrees of frequency (annually, semiannually, monthly, weekly, or daily). Finally *error correction* ensures that equipment not meeting the performance criteria or tolerance limit established for specific QC tests must be replaced or repaired to meet tolerance limits.

A QC *test procedure* outlines the steps taken in performing the QC test. These steps should be clear and precise and should be based on a format that is easy to follow. A potential format may include (a) name and purpose of the test, (b) equipment needed to perform the test, (c) testing frequency (how often the test should be conducted), (d) equipment setup and measurements to be recorded, (e) documentation of the test results, and (f) measures to correct unsatisfactory results [3].

Furthermore, it is essential that image quality criteria be a significant part of the QC program. These criteria are qualitative or quantitative. While qualitative criteria (pass-fail or uniform image without artifacts seen) are sometimes used to assess image quality, quantitative criteria should prevail and are obtained from scientific research. For example, when examining the exposure calibration of a computed radiography (CR) system, the AAPM recommends that "the exposure indicator response expressed in terms of exposure to 1 mR entrance exposure, the acceptance tolerance be $E_{\text{measured}} = 1 \pm 10\%$" [4].

Dose optimization in digital radiography will be described in detail in Chap. 12.

11.2.3 Parameters for QC Monitoring in Digital Radiography

QC for digital radiography [CR and flat panel digital radiography (FPDR)] has evolved from simple to more complex tests and test tools for use by radiology personnel to assure that the digital radiography equipment is working properly to ensure optimum image quality fall within the ALARA philosophy in radiation protection.

The American Association of Physicists in Medicine (AAPM) has recommended that several testing procedures for CR QC [4], using specific tools developed for CR QC. A few of these testing procedures, for example, include physical inspection of the imaging plate (IP), dark noise and uniformity, exposure indicator calibration, laser beam function, spatial accuracy, erasure thoroughness, aliasing/grid response, positioning, and collimation errors, to mention only a few. Examples of these tools include screen-contact wire mesh pattern, anti-scatter grid, calibrated ion chamber, densitometer for hard copy film evaluation (in cases where film is used), manufacturer-approved cleaning solutions and cloths, two metric-calibrated 30-cm steel rulers, and low-contrast phantom to mention only a few. Examples of CR QC tests will be highlighted later in the chapter.

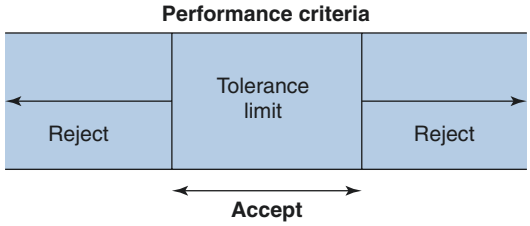

Fig. 11.1 A conceptualization of the notion of a tolerance limit and the consequences of exceeding the ± values that define the limit, in a QC program

11.2.4 Tolerance Limits or Acceptance Criteria

The AAPM has also established acceptance criteria or tolerance limits for the recommended digital radiography QC tests, in which both qualitative and quantitative criteria may be used to assess image quality. Consider Fig. 11.1 which conceptualizes the notion of tolerance limit and the consequences of exceeding the ± values that define the limit. For example, the tolerance for collimation of the X-ray beam should be ±2% of the source-to-image receptor distance (SID) [1]. For details on CR criteria and tests, the interested reader should refer to the AAPM Report No 93 [4], since it is not within the scope of this chapter to describe all of these tests.

11.3 Image Quality: Definition and Descriptors

11.3.1 Definition

In a paper on image quality, Zarb et al. noted that "image quality is difficult to define, since it is a nonspecific and subjective measure of the readability of a visual image" [5]. In its broadest sense, a "quality" image is one that makes accurate diagnosis possible. This is known as "diagnostic quality." The quality of any image can be described in terms of spatial resolution, contrast resolution, noise, detective quantum efficiency (DQE), and artifacts [6, 7].

11.3.2 Image Quality Descriptors

Image quality descriptors for CR and FPDR were described in detail in Chaps. 3 and 4, respectively. In review, the spatial resolution (sharpness) of a digital image is related to the size of the pixels in the image matrix. The smaller the pixel size, the better the spatial resolution of the image. Contrast resolution on the other hand is linked to the bit depth, which is the range of gray levels per pixel. An image with a bit depth of 8 will have 256 (2^8) shades of gray per pixel. Generally, a greater bit depth results in better contrast resolution. Noise can be discussed in terms of electronic noise (system noise) and quantum noise (quantum mottle). The quantum noise is determined by the number of X-ray photons (often referred to as the signal, "S") falling upon the detector to create the image. While low-exposure technique factors will produce few photons at the detector (i.e., less signal and more noise), higher exposure technique factors will generate more photons at the detector (i.e., more signal and less noise). The former will result in a noisy, or grainy, image that is generally poor, and the latter will produce a clearer image at the expense of increased dose to the patient. Finally, the DQE refers to the notion that the detector converts the input exposure (incident quanta) into a useful output image. The DQE is a measure of the efficiency and fidelity with which the detector can perform this task. Note that the DQE takes into consideration not only the signal-to-noise ratio (SNR) but also the system noise and, therefore, includes a measure of the amount of noise added. It is interesting to note that in his seminal article on CR physics, Rowlands [8], in elaborating on the clinical use of CR, stressed that "a primary criterion used by radiologists in comparing image quality…is the appearance of X-ray noise in the image or quantum mottle. Any visual signs of mottle make an image unacceptable" [8].

Furthermore, artifacts mask or mimic clinical features and constitute another aspect of diagnostic image quality. An image of exquisite quality that does not make its way to the physician for interpretation is of no value in the diagnostic process. QC must therefore address errors that affect both image quality and availability.

11.4 Understanding Processes and Errors in Digital Radiography

In this section, Dr. Charles Willis, PhD, DABR, Associate Professor, Department of Imaging Physics, The University of Texas MD Anderson Cancer Center, Houston, Texas, describes the elements of understanding processes and errors in digital radiography [9–12] (as mentioned in Sect. 11.1).

The ultimate product of the radiology operation is not the images, but rather the physician's interpretation (report) of the images. The quality of the report is beyond the scope of this chapter; however, it is important to recognize that QC activities in hospitals routinely monitor the quality of reporting, and national standards address the content and timeliness of communications between radiologists and clinicians. In order to determine where errors can occur and where QC measures should be instituted, it is helpful to consider each step in the process of performing a digital radiographic examination and producing the report.

11.4.1 Process Map for a DR Examination

A flowchart of the steps involved in performing a DR exam can be created, such as the one shown in Fig. 11.2. This flowchart is also known as a *process map*.

Constructing a process map is a way to understand the interrelationships between activities within the imaging operation. The individual steps can be very general or very detailed, depending on the purpose of the map. Maps for the same process differ depending on the specifics of local imaging operation. For example, an imaging facility that has an automated radiology information system (RIS) will have a different process map than a facility that uses manual methods for scheduling examinations. An imaging facility with a cassette-based DR system will have a different process map than a facility that uses an integrated DR system. For example, consider the process from arrival of the patient in the imaging department to release of the patient.

The process begins with the arrival of the patient in the imaging department. Steps 3 and 4 ensure that for a "walk-in" patient, the appropriate examination is scheduled in the RIS. This is an example of a QA activity, although no data was collected for QC purposes. Other QA activities are accomplished through interactions between the technologist and the patient, such as Step 7 where the exam is explained to the patient. At this time the technologist may verify that the patient is in fact the person identified in the exam request, that the anatomy to be examined matches the exam request, and other information about the patient such as pregnancy status, restricted motion, allergies if contrast is to be used, and appliances that may interfere with the radiographic projection or its development. A simple QC method is to include a checklist on the exam request that the technologist annotates to indicate that all these actions were taken. Step 12 is an inspection of the image for gross positioning errors that might require repeating a view. Step 14 is a more comprehensive review of the image at a workstation that is more capable of displaying subtle problems that might not be apparent at a preview station. The QC workstation in Step 14 might also have more sophisticated capabilities for modifying a substandard image without requiring a repeated exposure. The process map is designed so that images are released to the picture archiving and communications system (PACS) in Step 16, only after all have been reviewed and approved at the QC workstation.

Superimposed on the process map, eight boxes indicate places in the process where questions about QC arise. For example, before the cassettes are selected for the exam in Step 8, is each image receptor properly erased and ready to record another exposure? Is the correct patient identification associated with each cassette? Before positioning the patient in Step 9 for the examination, is the imaging system ready to make an appropriate exposure? Before the cassette is scanned in Step 11, is the scanner properly calibrated to develop the latent image in the receptor? When the images are viewed in Steps 12 and 14, how good is the fidelity of the image

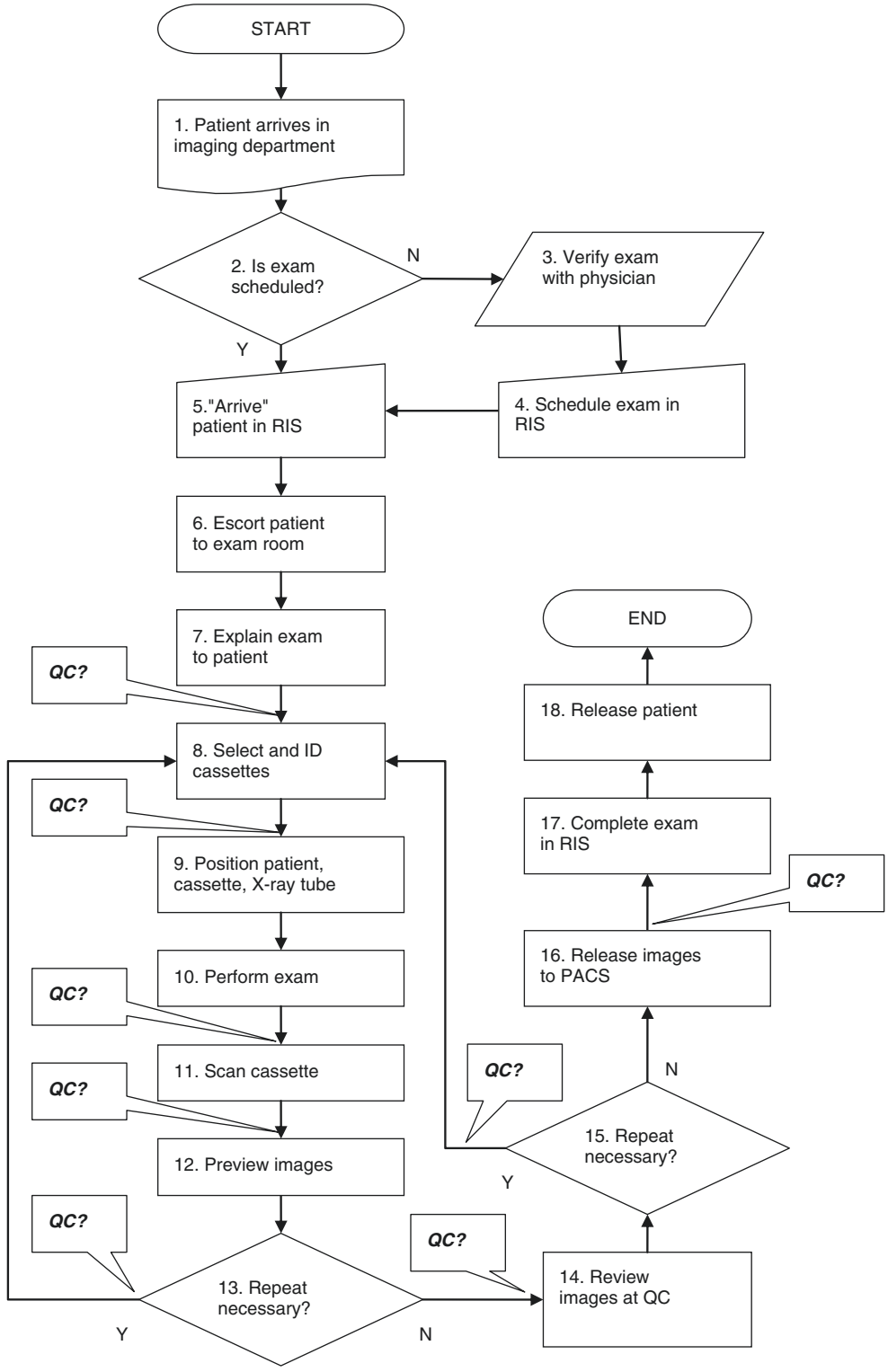

Fig. 11.2 Process map for DR examination. Each step of the process of performing a DR exam is depicted in the flowchart, with arrows indicating the sequence of steps. Places in the process where QC questions arise are annotated

display? Since the decision to repeat images in Steps 13 and 15 involves additional radiation exposure to the patient, how often does this occur and for what reasons? Once the images are released to the PACS in Step 16, did any or all of the images actually arrive at their intended destination? These questions can be answered by QC activities that are performed either as an integral part of the process map for each exam or periodically to avert or detect errors.

11.4.2 Errors in the Association of Demographic and Exam Information

In Step 8 of our process map, the technologist identifies both the patient and the exam to the imaging system. This usually occurs before the exam as shown in Fig. 11.2, but sometimes can be done after the examination is performed.

Every radiographic image has associated patient demographic and examination information, including patient name, patient date of birth or age, patient sex, a unique patient identification number, an exam accession number, the type of exam performed, the date and time of exam, and other information. In conventional screen-film radiography, the patient and exam information appears directly on the developed film. A window in the cassette allows the shadow of a printed version of the data to be flashed directly on the film. After the exam, when the film is developed, the data is visible. If the data is illegible, or incorrect, or no flash device is available, the data can be written onto the film with an indelible marker.

In DR, demographic and exam data accompanies the DR image in an electronic file header. This is called "metadata" and contains much more specific information about the image including how it was acquired and processed. The *Digital Imaging and Communications in Medicine* (*DICOM*) standard specifies the format and content of the header file. These fields, or tags, are named by group and element. For example, (0010,0020) is "Patient ID." A portion of metadata from a DR exam is shown in Fig. 11.3.

Where do all of these demographic and exam information come from? Demographic information may originate in the hospital information system (HIS) when the patient is registered. Demographic and some exam information may be present in the radiology information system (RIS) when the exam is scheduled or when the patient arrives. Some information may be provided in the exam request form. Some or all information may be recorded by the technologist at the time of the exam.

The simplest system is for the technologist to input all demographic and exam information into DR at the time of exam. Each DR system has some sort of registration device that allows direct manual input of demographic and exam information. Three basic drawbacks to this approach are the potential for human typographical errors, the fact that this is time-consuming, and that the data may already have been input by another person in the HIS or RIS introducing an inefficiency of "duplicate entry" and the possibility of discrepant information in the two automated systems.

Some vendors of cassette-based DR systems provide manual demographic association aids, as shown in Fig. 11.4. Special labels are affixed to the outside of the cassettes. The labels can be marked with a China marker or dry-erase marker to avoid confusion between patients and views. Where demographic and exam information exists in the RIS, an automated query to the RIS using the accession number as a key field can return all associated data to the DR acquisition station. The technologist still has to manually enter the accession number, which may be many digits, and verify that appropriate data is retrieved. The technologist may still have to manually enter some exam data. It is clear that RIS queries improve the accuracy of demographic and exam information, as well as reducing the technologist's time.

In imaging departments where the exam request is printed, a bar code of the accession number can also be printed. A bar code scanner can be added to the DR acquisition station as a substitute for the keyboard to reduce typographical errors. Some systems default to the previous patient's demographic information in the event of an unsuccessful query, so the technologist still must

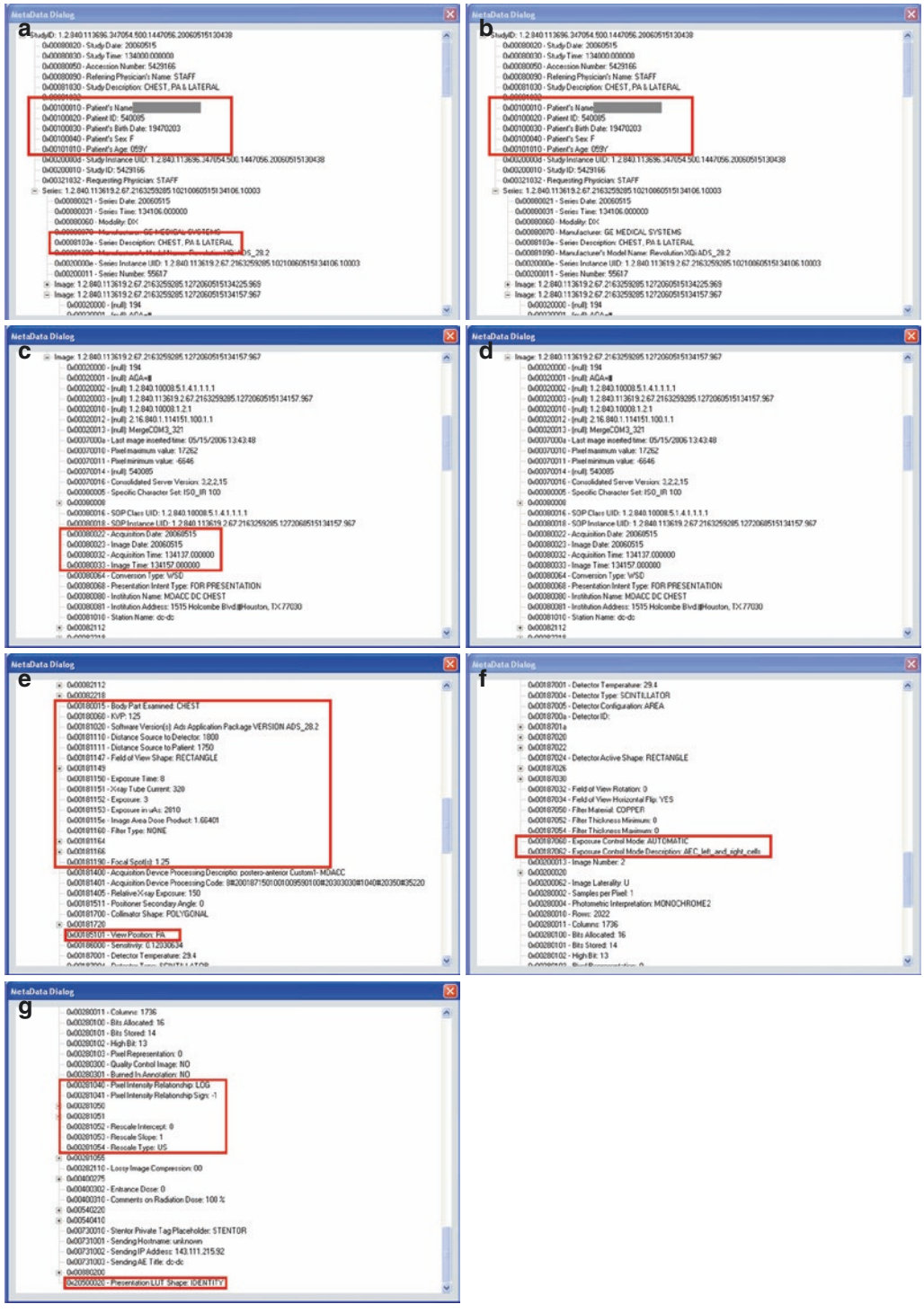

Fig. 11.3 Metadata from a DICOM image file. Each data element, or "tag," is identified by an eight-digit "group" and "element." Red outlines data concerning patient information and type of exam, orientation and time of exam, and details about how the image was acquired and how it should be displayed. This metadata is from a DICOM "DX" modality type. Some DR images are identified according to an older format for "CR" modality type

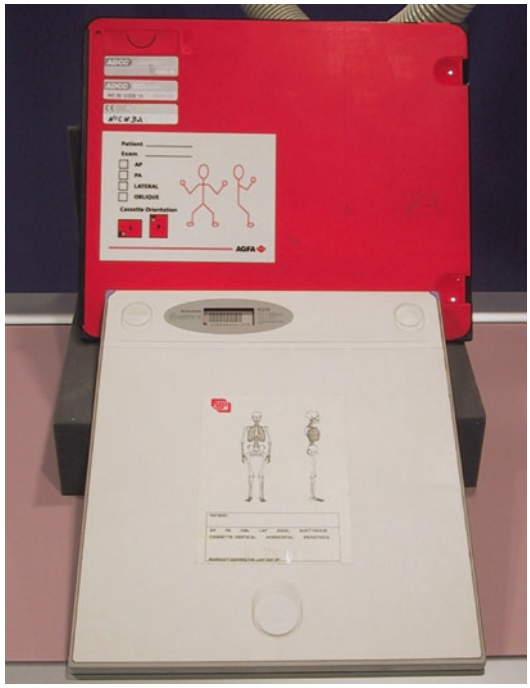

Fig. 11.4 Demographic association aids for cassette-based DR systems

verify that the appropriate data was retrieved. The technologist may still have to manually enter some exam data.

DICOM *Modality Worklist* (*MWL*) is a feature designed specifically to take advantage of information that already exists in the RIS. When an exam is scheduled for a particular imaging resource, the RIS makes a list of patients available to the DR acquisition station. The technologist only has to select the proper patient from the worklist in order to load all the demographic and exam information. When properly implemented, MWL makes a dramatic improvement in the accuracy of associations. Although MWL is a tremendous time-saver, it does not relieve technologists fully from vigilance over demographic and exam information. MWL has limitations, for example, in the case of an unscheduled exam, there may be a delay between scheduling the exam in the RIS and making it available to the acquisition station. Another limitation of MWL is resource reallocation, where, although the exam was scheduled for one imaging resource, for some reason, the exam must be conducted at another resource. A frequent practical example is when there is a delay in obtaining a post-void image, so that the next patient must be examined in another room. For this reason, often similar resources are pooled so that they share the same MWL; however, where all resources have been pooled, a search of the worklist is lengthy. Some patients have multiple exams scheduled for the same day, so the technologist must exercise care to select the proper exam from the MWL. MWL introduces a new type of error, that is, proper demographic and exam information that are associated with the wrong patient images. A simple technologist error in selecting the wrong patient or wrong examination from the worklist creates a condition that may be more difficult to detect than a typographical error. During an interruption in RIS services, MWL is not available, but the imaging operation may still need to continue. Therefore, a manual means to input and correct demographic and exam information must still be provided. MWL must be supported by both the DR acquisition station and the RIS and must be properly configured.

What are the consequences of mis-association of demographic and exam information and DR images? Incorrect information can cause the image to be unavailable for viewing. In a PACS, a mismatch between the image header and image database information from RIS generates what are known as "exceptions," "broken studies," or "orphans." These studies are usually sequestered and cannot be viewed by normal methods until the discrepancies are corrected. The clinical result is a "delay of diagnosis," which can have dramatic consequences for the outcome of the patient. Incorrect exam information can also affect image development. A chest image processed as a "skull" exam usually looks bad. A single exposure processed as a multiple exposure field exam is likely to be inappropriately developed. As mentioned previously in the discussion of MWL, mis-association also complicates error detection. Images assigned to a valid but wrong patient identifier are "lost." Images assigned to valid but wrong patient introduce the possibility of "misdiagnosis." Proliferation of digital

images within a PACS complicates correction. When a mis-association is discovered, how many other copies are floating around? Image caches on servers, workstations, web servers, gateways, remote sites, and archives all must be corrected.

11.4.3 Errors That Can Be Avoided by Periodic Testing

Returning to Fig. 11.2, only one of the QC questions that arose during analysis of the process map has been addressed. Most of the remaining questions concern the readiness of the DR system to acquire and display images. Because verifying every performance parameter before each exam would interfere with clinical operations, periodic measurements assure that the system is properly configured and calibrated and that performance has not degraded. This is precisely how the quality of other X-ray imaging equipment is managed. For example, the medical physicist performs annual tests of the X-ray generator to verify performance. These tests may be repeated following a service event. The technologist performs periodic tests, such as film processor sensitometry, to verify performance.

When performing the exam in Step 10, a number of potential errors can arise from improper configuration and calibration or degraded performance. These include inadequate erasure, improper compensation for nonuniform gain, incorrect gain adjustment, incorrect exposure factor selection, and artifacts. Artifacts arise from interference with the projected X-ray beam, image receptor defects, and interference with conversion of the captured projection into a digital image. These errors may be detected by inspecting the image at the acquisition station or at a QC workstation. Correction may be possible by digital image processing, or it may require a repeated exam. Most of these errors can be avoided by active QC measures, such as periodic testing. Some errors are the result of improper practice by the technologist, such as incorrect technique factor selection and artifacts from patient clothing or jewelry that interferes with the projected beam.

Unlike conventional screen-film image receptors, DR detectors are all inherently nonuniform. An example of an uncorrected image using an integrated DR receptor is shown in Fig. 11.5. Cassette-based systems have uniform surfaces to capture the radiographic projection, but the

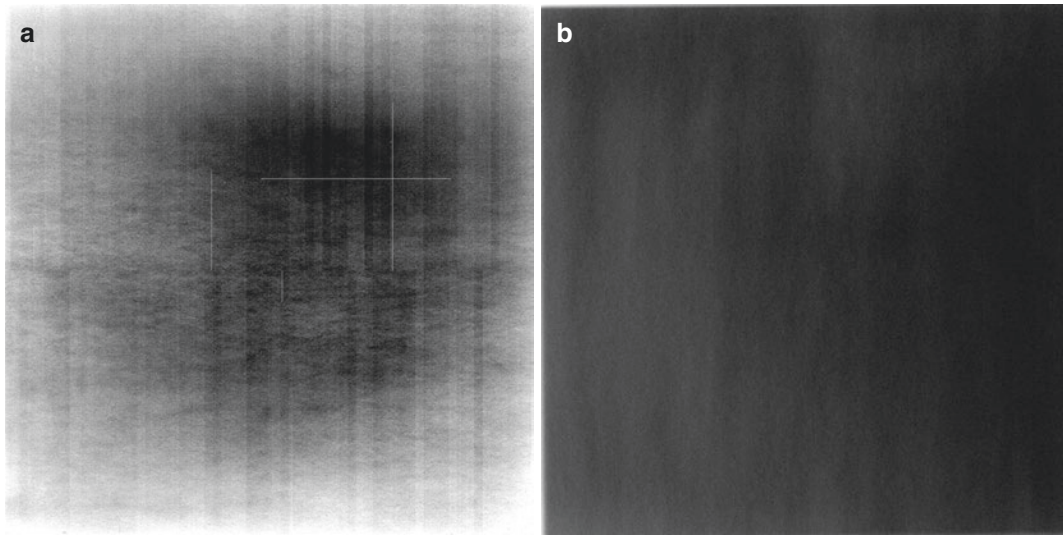

Fig. 11.5 Integrated DR receptor. (**a**) DR receptors are inherently nonuniform. (**b**) A uniformity correction must be calibrated on a periodic basis to compensate for differences in gain and artifacts such as dead pixels

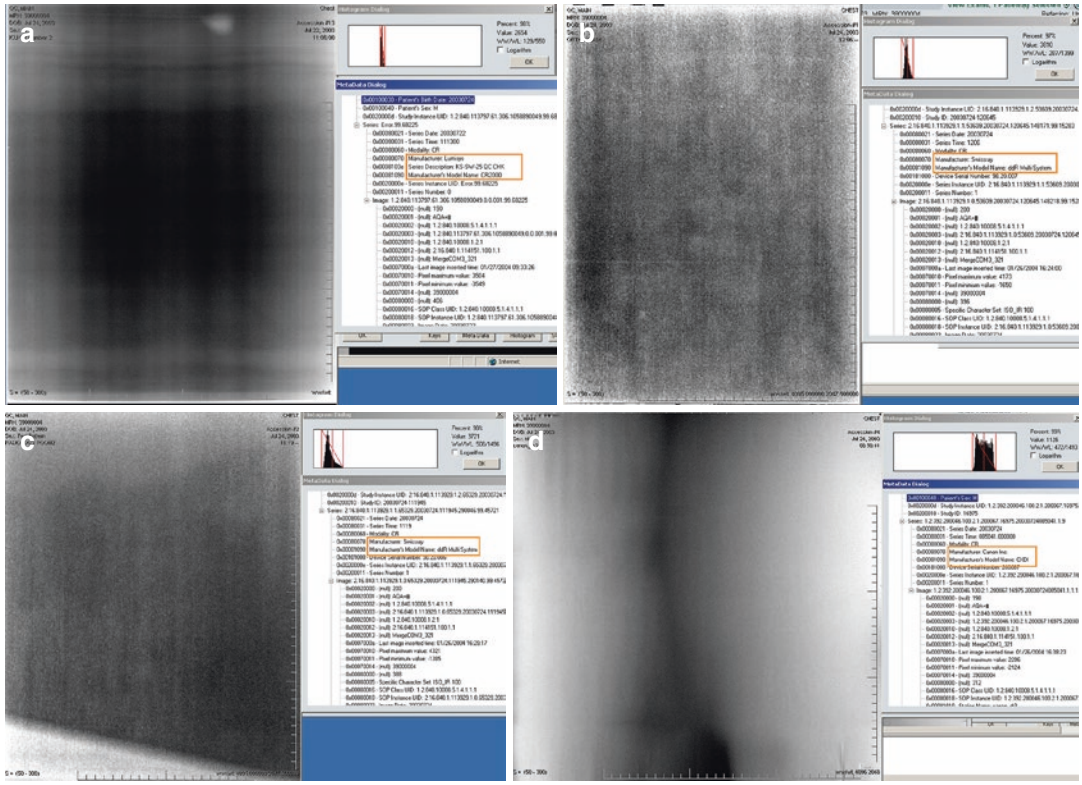

Fig. 11.6 Flat-field images on several DR systems. (**a**) LumiSys cassette-based DR. (**b**) and (**c**) Swissray integrated DR. (**d**) Canon integrated DR. Acceptable uniformity on (**a**) and (**b**) Unacceptable uniformity on (**c**) and (**d**)

efficiency of collection of stimulated emissions is highly nonuniform across the field of view. Both integrated and cassette-based DR systems require that a uniformity correction is applied to compensate for differences in gain across the receptor. This calibration for nonuniformity, also known as a *shading correction*, must be repeated on a periodic basis. The frequency of calibration depends on the specific DR device and ranges from daily to semiannually. A QC test for nonuniformity is easy to perform. All that is required is to expose the receptor to a uniform field of radiation, or *flat field*, which can be accomplished by using a large SID, such as 180 cm, opening the collimator to cover the receptor, and developing the image. Viewing the image reveals defects and improper uniformity calibration. During viewing, care should be taken in adjusting the contrast. At extremely high contrast, any DR image will appear nonuniform. This is illustrated in Fig. 11.6. Correction of nonuniformity improves both the contrast and sharpness of the DR image as shown in Fig. 11.7.

Another simple QC test helps answer the question raised in Step 8, concerning the readiness for the receptor to acquire another image. DR receptors are *reusable image media* (*RIM*), that is, they are not consumed in the process of acquiring an image. Instead, after the digital image is transferred, any signal remaining in the receptor is erased, so that the receptor can be used to acquire another image. If erasure is inadequate, the previous exposure is superimposed as a *ghost image* on the current image, as seen in Fig. 11.8. Developing an unexposed receptor, sometimes using a higher than normal gain setting, can reveal residual signal that has not been removed during erasure.

Inadequate erasure can arise from a variety of sources. For cassette-based systems, the duration

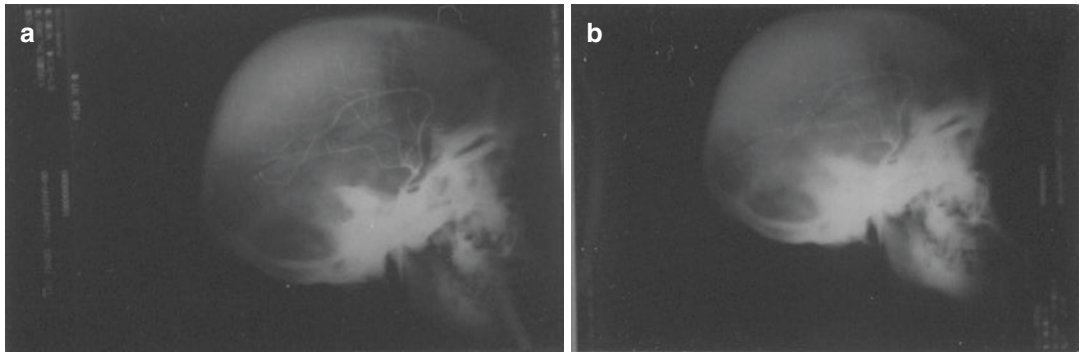

Fig. 11.7 Correction of nonuniformity on a cassette-based DR system. (**a**) The lateral view of a skull angiography phantom before nonuniformity correction. (**b**) After nonuniformity correction, contrast and sharpness are improved

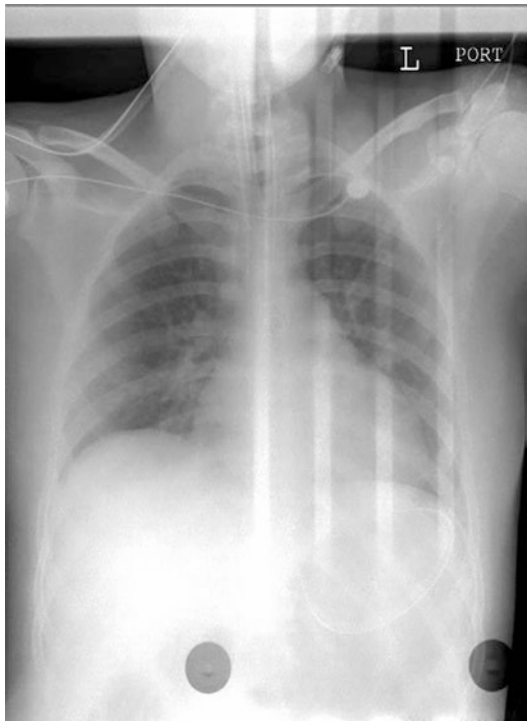

Fig. 11.8 Ghost image on cassette-based DR receptor

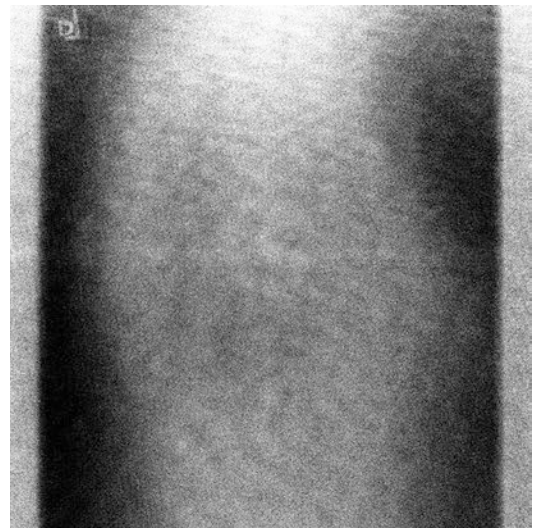

Fig. 11.9 Ghost image on indirect DR system

of erasure could be too short or the intensity of the erasure could be too low, perhaps from burned out erasure lamps. Scattered radiation from other examinations can fog a DR receptor in the same way that screen-film can be fogged. Fog can build up over time on DR receptors from environmental sources of radiation. Even with a proper erasure cycle, gross overexposure on the previous exam can leave residual signal. Ghost images can occur in both direct and indirect DR systems, although the specific mechanisms differ. Ghost images are usually associated with high-exposure levels with thick body parts and short times between exposures. Nonuniformity calibration when residual images are present can also cause ghost images that are reproduced many times, as seen in Fig. 11.9.

Because DR receptors are reused, artifacts arising from receptor defects, dust and debris that interfere with the collection of light for CR and indirect DR systems, and contaminants such as barium contrast will show up on all subsequent

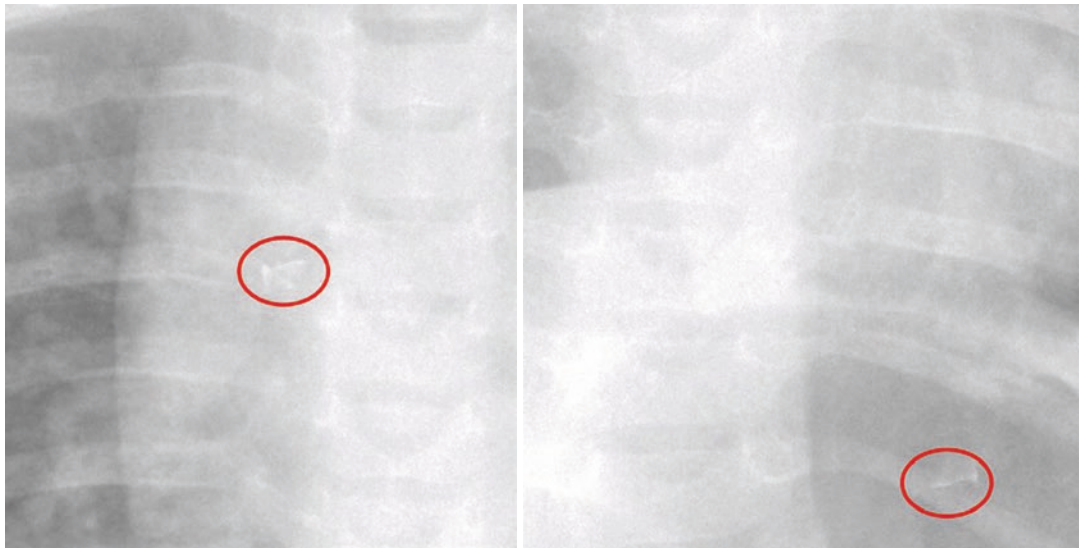

Fig. 11.10 Defect on image receptor. Artifact, circled in red, is visible in two clinical images of different orientation

images from that receptor until the problem is corrected, as shown in Fig. 11.10. Often the appearance of debris on the imaging receptor in a clinical image is difficult to distinguish from foreign bodies. With cassette-based systems, a visual inspection inside the cassette usually identifies the interfering material. With integrated systems, the receptor needs to be re-exposed to a uniform field to detect the presence of debris.

Every DR system has a gain adjustment which determines how much signal will be produced when the receptor is exposed to a specific amount of radiation. A flat field produced by specific exposure conditions of kVp, additional filtration, and measured radiation exposure to the receptor is needed to measure and to adjust gain. Each DR manufacturer has different conditions for the gain adjustment. The gain must be checked periodically and adjusted as required. Because the exposure conditions are critical for the gain adjustment, someone skilled in radiation dosimetry, such as a medical physicist, should be involved in the procedure.

In addition to checking gain, the same flat-field exposure serves to check the value of the DR *exposure indicator*. With screen-film radiography, the optical density (OD) of the developed film indicated how much radiation had been used in the examination. Because DR systems adjust their output signal to compensate for variations in exposure level, there is no convenient method for determining the amount of radiation that was used to produce the image. Almost every commercial DR system calculates some sort of exposure indicator. Exposure indicators have been standardized, and details of the International Electrotechnical Commission (IEC) and the AAPM of such standardized exposure indicator were described in detail in Chap. 5. It is important for technologists to understand the exposure indicator particular for DR systems in their imaging facility, because this is the only feedback provided to indicate overexposure or underexposure.

By acquiring the flat-field image according to the specific conditions of the manufacturer, it is possible to verify uniformity, absence of receptor defects, gain, and the value of the DR exposure indicator. In addition to periodic flat-field exposures, a list of proactive QC countermeasures can be compiled which emphasizes avoiding rather than correcting errors. When possible, all receptors could be erased at the start of each shift. The equipment and its environment could be cleaned periodically. A properly calibrated exposure indicator could be used to conduct periodic checks of the automatic exposure control (AEC).

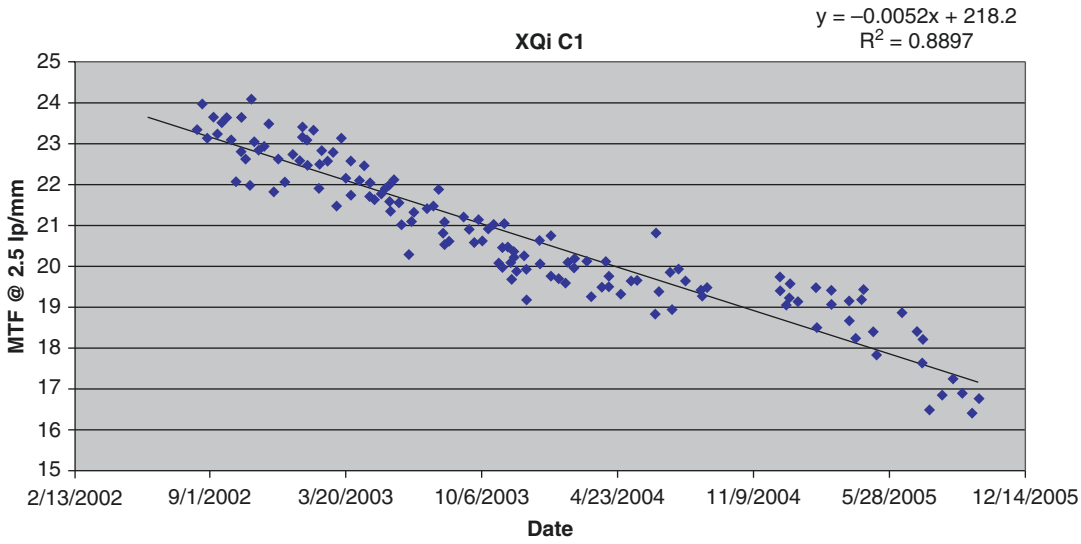

Fig. 11.11 Results of automated QC testing of a DR system. The graph shows a general decrease in the value of modulation transfer function (MTF) at 2.5 line pairs per mm calculated automatically in tests over several years. The minimum acceptable value established by the manufacturer is 17%. The detector required replacement

Likewise, the exposure indicator is crucial for validating a technique guide for manual exposure factor selection. Thorough AT could be conducted for all new receptors. Checks of DR system performance could be repeated immediately following service events and software upgrades.

The AAPM Task Group 10 worked with CR vendors to specify AT and periodic QC tests for CR. Although AT and QC for other DR systems have not been defined at this writing, the CR tests constitute a reasonable starting point. DR manufacturers have developed a variety of phantoms and automated tests to assist us in assessing performance on a periodic basis. The interested reader should refer to the various manufacturers for their specific test tools.

Figure 11.11 demonstrates the gradual degradation of sharpness in a DR detector indicated by automated QC tests. It is also interesting to note the large amount of variation in the data.

11.4.4 Errors in Performing the Examination

The one test that can and should be done for every DR image is a visual assessment of image quality before release. In Steps 12 and 14 of the process map, each image is reviewed to determine whether image quality is satisfactory or whether the projection should be repeated. Inspection of the images reveals errors that might be made by the technologist positioning the patient in the radiation field and performing the examination in Step 9, such as mispositioning, patient motion, incorrect radiographic technique selection, poor inspiration, improper collimation, incorrect alignment of X-ray beam and grid, wrong exam performed, and double exposure. There is little that digital image processing can do to improve these substandard images. Correction of these errors is likely to require repeated views.

Other errors can arise when the image receptor renders the captured projection for viewing, which would be during Step 11 in the process map. Potential errors include incorrect recognition of the exposure field resulting in incorrect determination of the values of interest (VOI) in the image, incorrect histogram rescaling, incorrect grayscale rendition, incorrect edge restoration, inappropriate noise reduction, and incorrect reorientation for the projected anatomy. Correction of these errors is usually possible at the QC workstation without a repeated exam.

Sometimes improper operation of the DR system is the underlying cause of a failure to detect the radiation field. When the technologist employs nonparallel collimation, multiple exposure fields on a single cassette-based receptor, and poor centering of the anatomy on the receptor or violates collimation rules, and in some instances where an implant or auxiliary shielding overlies one or more collimation boundaries, the algorithm that detects the edges of the radiation field is confounded. The result is inappropriate histogram analysis and incorrect rescaling of the image for display.

Since a key aspect of QC is a visual inspection of the DR image, it is important to consider the capabilities of the QC workstation where we are making decisions in Step 14 about repeating or correcting images. In some DR systems, these functions are incorporated into the acquisition station or identification station that is used for performing the exam or developing the DR image. Traditionally, QC workstations allow the technologist to modify image processing, modify demographic and exam information, imprint demographic overlays, add annotations, apply borders or shadow masks, flip and rotate images, increase magnification, conjoin images to make composite images for scoliosis and full leg exams, modify the sequence of views, verify exposure indicator, and select images for archive or transmission to PACS. The QC workstation usually includes the capability of deleting images.

Bad electronic images can disappear without a trace: bad films can also disappear, but leave a signature. Technologist supervisors learn quickly to count how many films are in the box at the start of shift, count again at the end of shift, and compare against the number of views produced by all exams during the shift. Some manufacturers have begun to incorporate in their QC workstations the capability to collect data on repeated and rejected images for reject analysis. If this feature is not available on the QC workstation, an alternate method will need to be developed for documenting the number and causes of rejected or repeated images.

11.4.4.1 Reject Analysis

Reject analysis is a time-honored method for assessing and improving quality of imaging operations. A reject analysis involves an examination for it is not enough to simply collect the data: the data must be analyzed to reveal recurring causes, the results need to be reported to management and staff, and action needs to be taken based on the results in order to effect any improvement. Management should implement remedial training as indicated. It is worthwhile to share the results with the DR vendor, because they may be able to provide advice or modify image processing settings for problematic exams.

Early adopters of DR technology claimed few or even no rejected images. Considering the number of technical errors that can be made in projection radiography that are independent of the acquisition technology, an unrealistically low repeat rate implies that the standards for acceptable quality were also low. An early study by Dr. Willis [10] provided results of DR reject analysis from one hospital that are shown in Table 11.1. These are fairly typical in cause and frequency to the results of reject analysis in conventional screen-film operations. Positioning errors are by far the most frequent cause of repeated exams. Later studies on reject analysis were conducted by Bjorn et al. [13], Khafaji et al. [14], and Andersen et al. [15].

Table 11.1 Reject analysis for 1 month from a hospital using DR

Reason	Number	%
Positioning	489	46.9%
Overexposed	122	11.7%
Underexposed	105	10.1%
Reprinted	89	8.5%
Motion	57	5.5%
Over-collimated	40	3.8%
Artifact	23	2.2%
No exposure	21	2.0%
Double exposed	17	1.6%
No marker	10	1.0%
Marker over part	8	0.8%
Other	8	0.8%
Total	*1043*	*100.0%*

Bjorn et al. [13] study showed that the reject rate was comparable to that of film-screen imaging systems. While the overall deletion rate was 11%, 51.3% were deleted because of positioning errors, and 31% were attributed to centering errors. Finally the authors reported that those examinations with the highest deletion rates were knees (20.6%), hips (18.5%), and ankles (13.8%). The study by Khafaji et al. [14] showed a rejection rate of 15%, with positioning errors being one of the major reasons for rejection, followed by artifacts (28.5%) and motion (17.1%). Furthermore, examinations of the pelvis, abdomen, spine, and knees showed higher rejection rates than the average. Finally, the results of the study by Andersen et al. [15] showed an overall rejection rate of 12%. While 77% of the rejected images were due to positioning errors, the examinations with the highest rejection rates were knees, shoulders, and wrist.

One of the tasks of the technologist after the DR image is acquired is to determine whether the image is underexposed or overexposed. You can see from Table 11.1 that only a small proportion of repeated images were caused by improper exposure. This is normal for an imaging operation that has appropriate targets for DR exposure. As mentioned previously, because of automatic rescaling of the digital image, the "darkness" or "lightness" of the DR image is not a reliable way to assess exposure factor selection. When few X-rays are used to produce a DR image, the image appears noisy. As more X-rays are used to produce the DR image, the image appears less and less noisy. Technologists are quick to learn that physicians are less likely to reject overexposed DR images than underexposed DR images. Since repeating exams is an unpleasant experience, there is a documented tendency for technologists to increase the exposure factors in DR exams. The exposure is increased by using higher technique factors than indicated on the technique guide for manual exams and by using plus density settings on AEC exams. This tendency is known as *exposure factor creep*.

The good news is that QC programs that track DR exposure indicators are effective in controlling exposure factor creep. Figure 11.12 shows

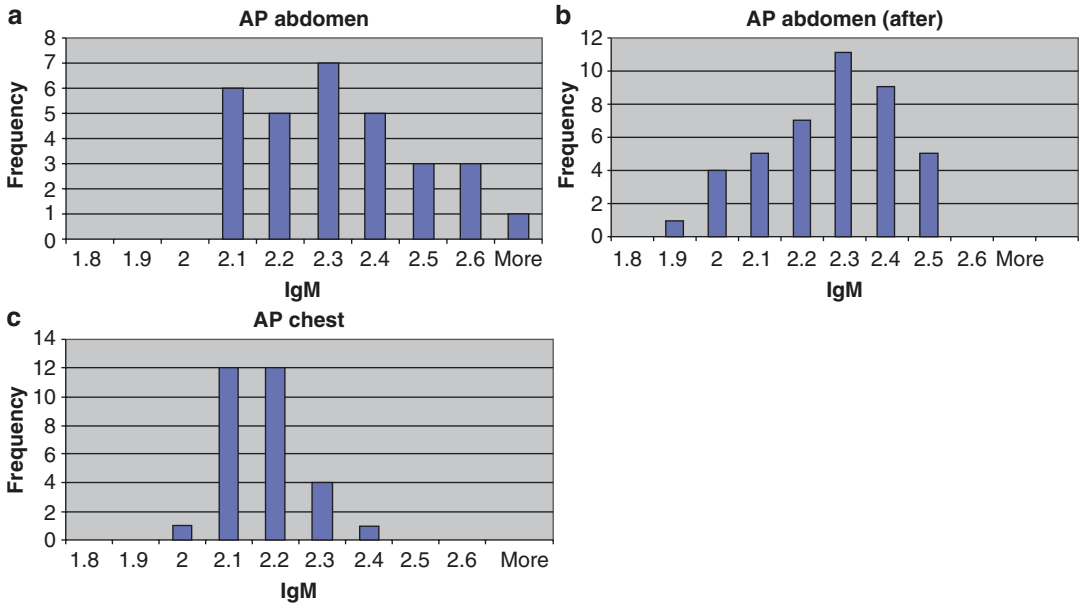

Fig. 11.12 Distribution of DR exposure indicator values for specific radiographic views. The first and second histograms show the distribution of lgM values for AP Abdomen exams. The target value was 2.2 ± 0.3. Even though the average value was similar after QC efforts, the distribution is clearly shifted toward lower exposures with none exceeding the maximum acceptable value of 2.5. The third histogram shows a much narrower distribution of values for AP chest exams

exposure indicator values observed for a specific exam and view. The value of the DR exposure indicator is usually visible at the QC workstation. Some vendors have developed software that facilitates collecting exposure indicator data. If your DR device does not have this sort of software, an alternate method for collecting this data will need to be developed.

Automatic rescaling contributes to a common misconception that in DR exposure factor selection is irrelevant to the appearance of the image. This myth arises from misunderstanding about the differences between *latitude*, *exposure latitude*, and *range of adjustment* in DR. Latitude is used by imaging scientists and engineers to mean the ratio between the highest detectable radiation level and the lowest detectable radiation level for an image receptor. For DR receptors this may be a huge value, such as 10,000! Exposure latitude is used by technologists to mean the ratio of the highest exposure factor selection to the lowest exposure factor selection on the X-ray generator that will produce an acceptable radiographic image. Even in DR, exposure latitude is a much smaller value than image receptor latitude. Exposure latitude is reduced by the range of radiation levels that are produced by range of tissue densities and thicknesses of the anatomy included in the radiographic projection. The ratio between the highest radiation levels in a projection, such as the areas near the skin or through the lung apices, and the lowest levels, such as the bones, is about 100. Even if the DR receptor had a latitude of 10,000, the exposure technique could only increase or decrease by a factor of 10 from the center of the receptor dynamic range and still acquire the entire range of radiation levels in the projection. That means that in this case the DR exposure latitude is only 100. This is still a large number compared to screen-film exposure latitude which is about 2.

The exposure latitude is further limited by the range of adjustment of the DR system. Each DR system can only compensate for a limited amount of overexposure and underexposure for a single image. For example, one DR system can only display a range of 400 in radiation exposure once the settings for a specific acquisition have been determined by the operator, even though its receptor latitude is 10,000. This limits the exposure latitude to a factor of 4, which is still twice as good as screen-film.

The consequences of exceeding the limit of adjustment by underexposure and overexposure in DR have been traditionally neglected in the technical literature. Vendors who had been reluctant to acknowledge any limitations on their systems have begun to issue some cautions about gross overexposure. Underexposure in DR causes an increase in quantum mottle and a loss of contrast in dense features, as illustrated in Fig. 11.13. It constitutes about 9% of all rejects. Overexposure can cause a similar loss of contrast in skin and dense features, as shown in Fig. 11.14, and makes up about 5% of all rejects. Both underexposure and overexposure can confuse the exposure data recognition software, as shown in Fig. 11.15. This image processing software decides which values in the image are of interest (VOI) and rescales the image accordingly.

If technologists are expected to make decisions about whether to repeat exams in Step 13 and Step 15 and whether to modify the appearance of the image before releasing to PACS, then the fidelity of the DR image that is displayed on a preview station, acquisition station, or QC workstation is a matter of concern. If the appearance of the image to the technologist differs from appearance to the physician, the technologist should not be allowed to modify the image. In order to ensure that the appearance of the image at the QC workstation matches the appearance at the physician's PACS workstation, both displays must be similarly calibrated. DICOM specifies a standard *grayscale display function* (*GSDF*) which helps in matching the appearance of the image on different electronic displays and helps to equalize the human observer's perception of contrast in light and dark areas of the image. The same function can be applied to laser film printers that can serve as the output device for DR in some imaging facilities. Some electronic displays automatically apply the GSDF and some automatically compensate variations in ambient illumination and for changes in display performance.

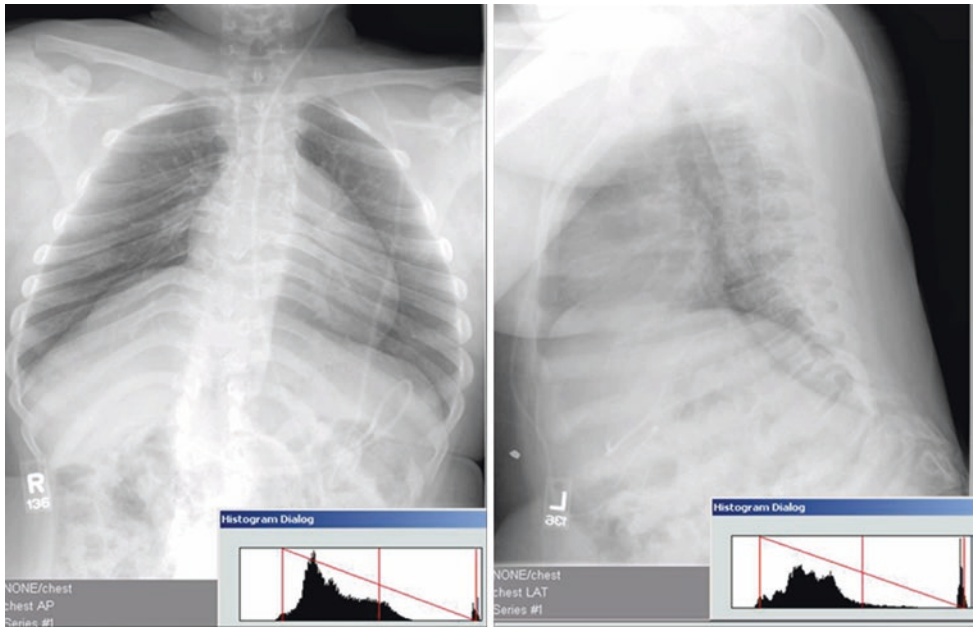

Fig. 11.13 Underexposure in DR. Both views are underexposed. Note the grainy appearance and loss of contrast in dense features

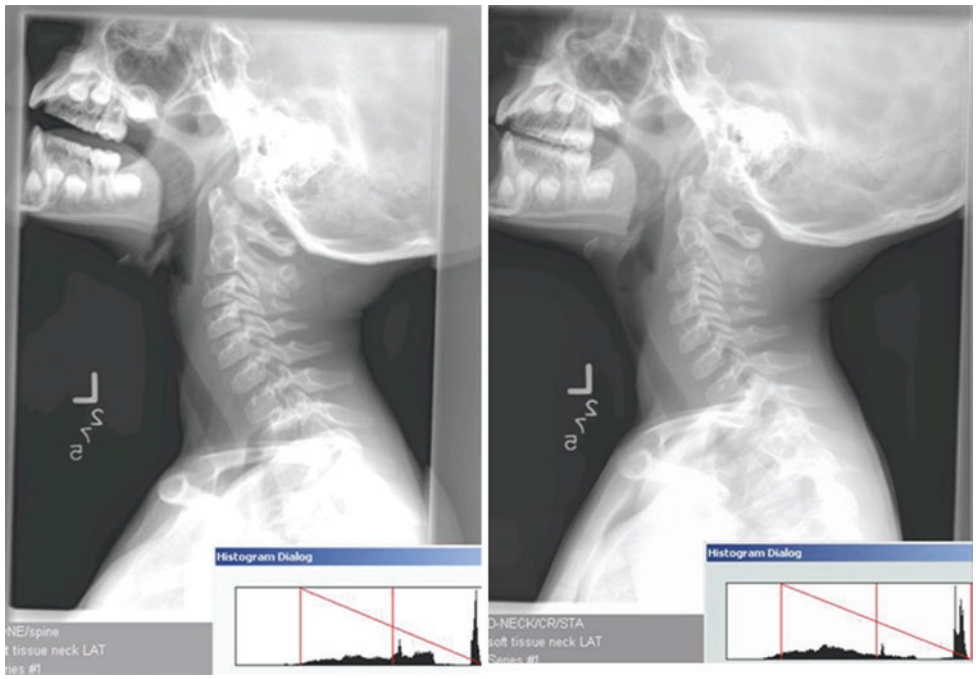

Fig. 11.14 Overexposure in DR. The view on the left was overexposed. The exposure indicator was at the high end of the target range. The view on the right was repeated at one-half the mAs of the first image. Note the loss of contrast in the skin and dense areas in the overexposed image

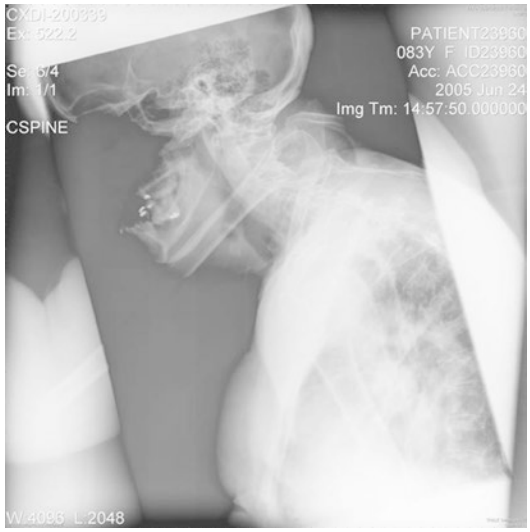

Fig. 11.15 Exposure data recognition failure in DR. The image was grossly overexposed by a factor of almost 50. The detector was saturated. This is the result of exceeding the limit of adjustment

AAPM Task Group 18 has defined QC tests for electronic displays intended for primary interpretation. These QC tests should be repeated periodically to verify proper display performance. The frequency of testing depends on the specific display technology.

Because images appear at the QC workstation before they are available on PACS, physicians who have access to the QC workstation are likely to look at this display and to make decisions regarding the treatment of the patient. Imaging departments are wise to regard this as a display for diagnostic purposes and maintain it accordingly.

One of the frequent modifications of DR images before release to PACS is the annotation of the images to indicate laterality. This is needed when the lead marker is inadvertently excluded from the radiation field. Unfortunately, routine dependence on annotation post-exam is prone to human error and should be discouraged. Just like conventional screen-film radiography, the lead marker should be included in radiation field as a contemporaneous record in the DR image.

11.4.5 Errors in Delivery of the Images

When the acquisition station transfers the DR image to the PACS archive, there are several potential errors that can occur. For example, the transmission can be interrupted without any notification of the sender. The image can arrive safely only to be deleted from local cache. The image can arrive safely except that critical information present in the original image can be omitted from transmitted image, such as the exposure indicator, processing parameters, shutters, and annotations. This information may actually be present in the transmitted image but may not be displayable by the PACS viewer.

Unlike conventional screen-film radiography, in DR the technologist or supervisor must accept responsibility for appropriate delivery of all images to the physician. Processes must be in place to verify that all exams performed and all images acquired reach their intended destinations. In this regard, an image count of two does not necessarily mean that both the PA and LAT views arrived. Processes must be in place to correct delivery errors when detected.

Ultimately, the DR image is displayed for viewing by a physician. Regardless of whether the physician is viewing the image in soft copy, on an electronic display, or hard copy, a laser-printed film trans-illuminated on a light box, there are a number of potential errors that affect image quality. The display device may have an incorrect GSDF calibration. The display may have an inadequate pixel matrix to display the full resolution of the DR image. This can either result in spatial resolution that is inadequate to visualize fine clinical details or moire patterns (interference patterns) with periodic features in the DR image such as stationary grid lines. The value of the image can be compromised by incorrect or missing demographics or annotations, inadequate viewing conditions such as encountered environments with high ambient illumination and glare, and errors not detected and corrected by previous QC. It is important for the radiologist to feedback information about the ultimate quality of the images to the technologists who are producing the images.

11.5 Selected QC Tests for CR: Qualitative Criteria

As mentioned above, the AAPM [4] has identified testing procedures for CR systems including, for example, physical inspection of the imaging plate (IP), dark noise and uniformity, exposure indicator calibration, laser beam function, spatial accuracy, erasure thoroughness, aliasing/grid response, positioning, and collimation errors. This section presents a brief outline of QC tests for dark noise, CR screen test = uniformity, spatial accuracy, and erasure thoroughness.

11.5.1 TEST 1: Dark Noise

Purpose: To assess the level of noise present in the system (intrinsic noise).

Exposure Condition

- No exposure. Erase a single screen and read it without exposing it.

Process in the Image Reader
Qualitative Criterion for Acceptance

- Uniform image without artifacts as shown in Fig. 11.16.

11.5.2 TEST 2: CR Imaging Plate Test for Uniformity

Purpose: To assess the uniformity of the recorded signal from a uniformly exposed imaging plate (a nonuniform response could affect clinical image quality).

Exposure Condition

- Expose imaging plate using appropriate exposure factors.

Process in the Image Reader
Qualitative Criterion for Acceptance

- Uniform image without artifacts, as shown in Fig. 11.17.

11.5.3 TEST 3: Spatial Accuracy

Purpose: To check that there is no spatial distortion in the image.

Exposure Condition

- Place a regular wire mesh screen-film contact test tool over the CR imaging plate.
- Expose the imaging plate to appropriate exposure technique.

Fig. 11.16 Qualitative criterion for acceptance for the CR IP QC test for dark noise is a uniform image without artifacts

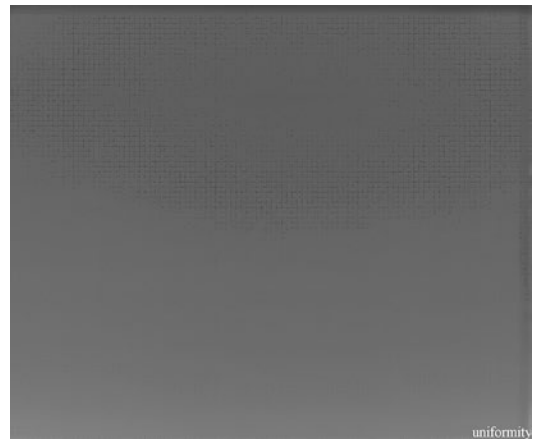

Fig. 11.17 Qualitative criterion for acceptance for the CR IP QC test for uniformity is a uniform image without artifacts

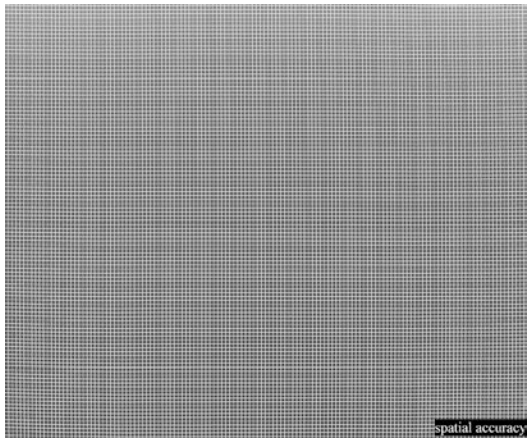

Fig. 11.18 Qualitative criterion for acceptance for the CR IP QC test for spatial accuracy shows that the grid pattern spacing of the wire mesh should be uniform without any distortion across the CR image

Process in the Image Reader
Qualitative Criterion for Acceptance

- The grid pattern spacing of the wire mesh should be uniform without any distortion across the CR image, as shown in Fig. 11.18.

11.5.4 TEST 4: Erasure Thoroughness

Purpose: To test that minimal residual signal (ghosting) on a CR imaging plate after readout and exposure.

Exposure Condition

- Place a step wedge at the center of a 14 × 17 CR IP, and expose using appropriate exposure technique (to image the step wedge).

Process in the image reader

- *Re-expose* the same IP a second time without the step wedge using the appropriate exposure technique.
- Collimate in by about 5 cm on each side of the CR IP.

Process in the Image Reader
Qualitative Criterion for Acceptance

- The absence of a ghost image of the step wedge from the first exposure in the re-exposed image, as shown in Fig. 11.19.

11.6 Ongoing Quality Control

It is quite clear that ongoing QC is an essential element and process of any digital imaging department, and as such the AAPM has provided details of such activities, as described by Jones et al. [16]. The authors also identify three major themes of ongoing QC: reject image analysis, exposure analysis, and artifact identification. They stress the continued need for reject image analysis since repeated and rejected images influence the dose to the patient and the "inefficiency in the imaging operation owing to wasted time and resources" [16]. The AAPM suggests that an acceptable reject rate should fall between 6% and 10% [16].The rejection rates mentioned above in the subsection on reject rates would indicate that there is a need for improvement of radiographic practice and a need for the optimization of radiation protection. Furthermore, the AAPM suggests that the types of data "required for a functional reject analysis program [(RAP) ("required")] would make a RAP simpler and more useful ("optional")."

While examples of such required data would be accession number and acquisition station, date and time of the examination, body parts and view, exposure indicators, reject category, and technologist identification, examples of optional data would include reject comments, name of the technologist and exposure technique factors used for the examination, and a thumbnail image (Table 11.2) [16].

Exposure analysis is vital to digital radiography QC procedures because of the phenomenon of dose creep due to the wide dynamic range of digital radiography systems, as described in detail in Chaps. 3 and 4 and as outlined by Dr. Charles Willis in the above section. The use of exposure indicators (EIs) is intended to address the problem of dose creep, since they provide direct feedback to the technologist as to the use of correct exposure techniques for the examina-

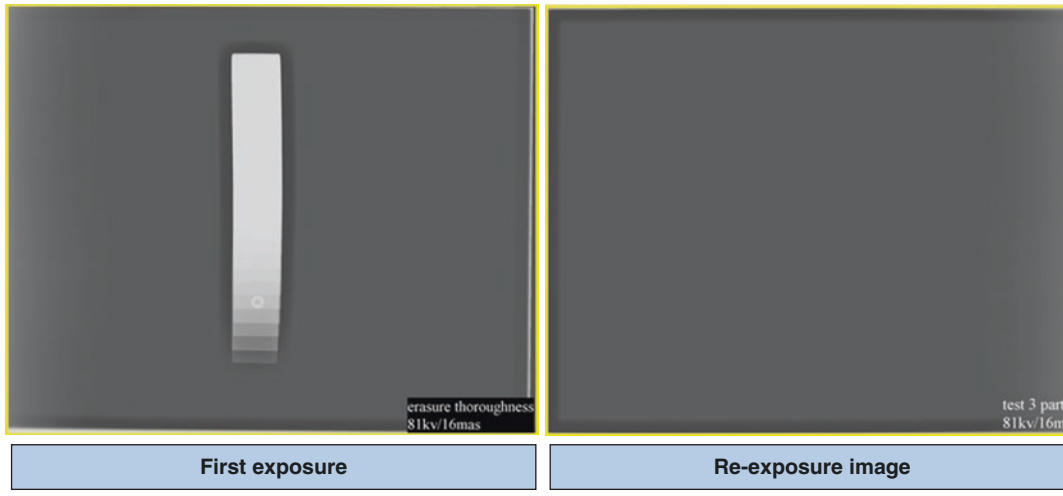

Fig. 11.19 Qualitative criterion for acceptance for the CR IP QC test for erasure thoroughness shows the absence of a ghost image of the step wedge from the first exposure in the re-exposed image

Table 11.2 Data stored for rejected image analysis (From Jones AK, Heintz P, Geiser W, Goldman L, Jerjian K, Martin M, et al. Ongoing quality control in digital radiography: Report of AAPM Imaging Physics Committee Task Group 151. Medical Physics 2015; 42 (11): 6658–6670. Reproduced by permission of Wiley & Sons)

Field	Function	Required/optional
Acquisition station/digitizer	Can identify specific stations with problems	Required
Accession number	Links study to technologist through RIS	Required
Exam date and time	Allows temporal sorting of data	Required
Body part	Allows sorting of data by body part	Required
View	Allows sorting, of data by view	Required
Exposure indicators (EI)[a]	Allows exposure analysis/troubleshooting	Required
Reject category	Allows reject analysis	Required
Technologist ID	Alternative method of linking technologist and study	Required[b]
Reject comments	Further clarifies reason for rejection—free field	Optional
Technologist name	Allows sorting of data by technologist name	Optional
Technique factors	Troubleshooting	Optional
Thumbnail image	QC of reason for rejection	Optional

[a]The target EI and DI should also be included, if available
[b]Optional if separate user names are provided for each technologist who uses the system

tion. In a study of the use of the EI as a dose management strategy, Seeram et al. [17] showed selecting optimized mAs reduced dose by 36% compared with the vendor's recommended mAs (dose) values and subsequently concluded that optimizing the mAs and associated EIs can be an effective dose management strategy [17]. Details of the use of the EI as a dose management strategy will be discussed in Chap. 12 which deals with dose optimization in digital radiography.

Finally, artifact identification is a critical task of both technologists and radiologists. Since the technologist evaluates images for image quality before sending them to the PACS for the radiologists to interpret, it is vital that the technologist has a firm understanding of artifact identification. To assist the technologist in this regard, Jones et al. [16] suggest the use of what they refer to as a "fault tree" as shown in Fig. 11.20. The fault tree not only identifies the actors but actions that should be taken should an artifact be present on

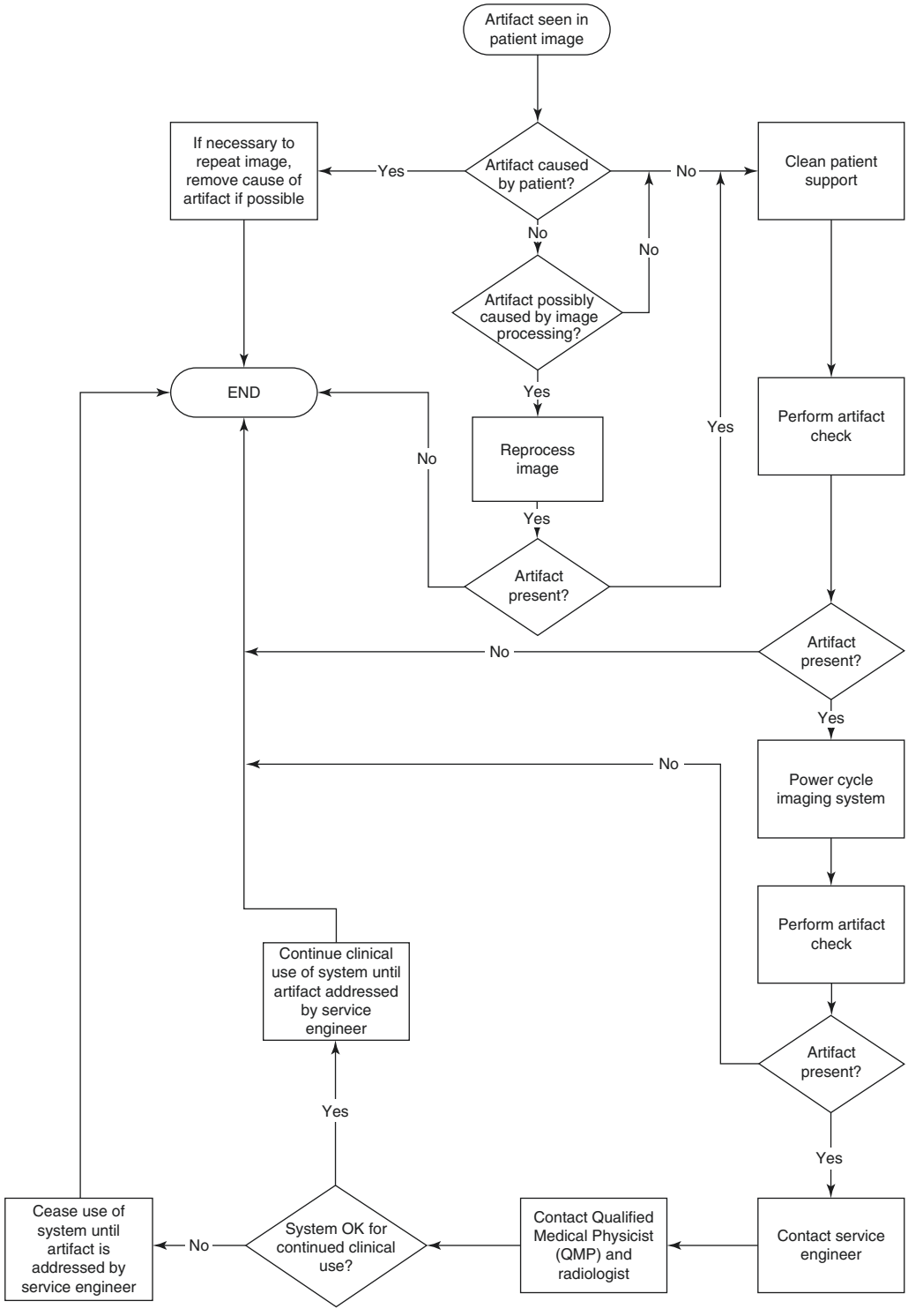

Fig. 11.20 Fault tree for artifact troubleshooting. (From Jones AK, Heintz P, Geiser W, Goldman L, Jerjian K, Martin M, et al. Ongoing quality control in digital radiography: Report of AAPM Imaging Physics Committee Task Group 151. Medical Physics 2015; 42 (11): 6658–6670. Reproduced by permission of Wiley & Sons)

the image [16]. Furthermore the authors also provide a "protocol for performing artifact check." Interested readers should refer to the article by Jones et al. [16] for further details.

11.7 Responsibilities for DR QC

In order to be effective, it is important to involve all local resources in a team approach to the QC effort. Active participation by the radiologist is an absolute requirement. The radiologist has the ultimate responsibility for quality of images. Within the imaging operation, resources and priorities are controlled by the radiologists. Radiologists set the standard: hospital staff can only produce the lowest level of quality that is acceptable to the radiologists. The radiologist must demand accountability for image quality and availability and must enthusiastically support the QC effort. Other members of the QC team will follow the example set by the radiologist.

11.7.1 The QC Team

The ACR, AAPM, and SIIM in their technical standard for electronic practice of medical imaging identify and describe at least five individuals who should play a significant role in QA/QA programs including the physician, a qualified medical physicist (QMP), registered radiologist assistant, radiologic technologist, and the imaging informatics professional [18].

As described by Willis [19], the radiology administrator is also a key player in QC. The administrator is responsible for efficiency of imaging operations, and QC is a means of improving efficiency. The lead radiologic technologist is the first-line supervisor of QC operations. The clinical engineer is responsible for equipment life cycle management and is therefore intimately involved in calibrations and service. Clinical engineers are familiar with DR acquisition devices, X-ray generators, AEC devices, and electronic displays. Medical informatics is a relatively new discipline with an important role in modern imaging operations. A medical imaging specialist can assist with collection of QC data and troubleshooting of QC issues that arise from interactions between the RIS and the DR image acquisition stations, with networking, and with workstation configurations. The QMP is uniquely qualified to interpret the meaning of QC results in the context of clinical practice.

11.7.2 Radiologist Feedback

The pivotal role of the radiologist in providing feedback on QC merits some elaboration. Incidental comments and guidance regarding image quality is valuable. However, routine, systematic, documented assessment of image quality is much more valuable for the technical staff. In the film-based radiology department, periodically the radiologist would conduct a critique of a sample of films for interpretation. It is possible to institute a similar process in the digital department. In order to minimize interference with the radiologist workflow, image critiques can be categorized into codes that are transcribed into report for subsequent automated extraction. Critiques should include availability as well as quality items. A system such as this documents causes and frequency of substandard imaging and tracks improvement. This is a mechanism for establishing responsibility for the quality of service provided by the imaging operation. Figure 11.21 shows images that were released to the radiologist despite an active QC program.

11.7.3 Defining QC Responsibilities

Having considered specific QC activities that can help monitor and improve image quality, it is useful to construct a table that identifies each QC task and the party responsible for performing the task. This table is referred to as a *task allocation matrix*. Additional information can be added, such as the frequency that the task is performed and a reference for the procedure to be followed. An example is included as Table 11.3.

As mentioned above, the ACR, AAPM, and SIIM have identified the imaging informatics

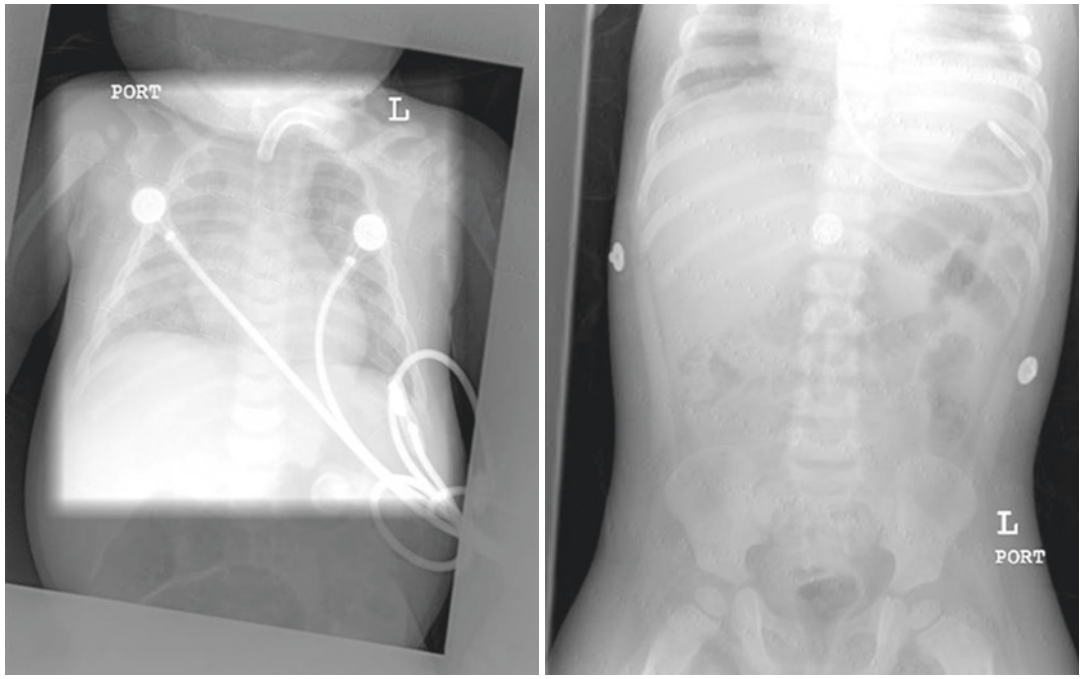

Fig. 11.21 Substandard DR images released to the radiologist for interpretation. The radiologist reported a problem after receiving the image on the left. Note the periodic diagonal artifact going across the image from upper left to lower right. Later, the radiologist received the image on the right with a similar artifact, because no action was taken at the first report

Table 11.3 QC task allocation matrix

Task	Responsibility	Frequency
Verify patient ID and exam info	Technologist	Each exam
Verify patient positioning	Technologist	Each view
Verify image quality—release or repeat	Lead technologist	Each image
Verify exam in PACS	Lead technologist	Each exam
Reconcile patient data/image counts in PACS	Informatics	Incidental
Report substandard images	Radiologist	Incidental
Erase cassette-based image receptors	Technologist	Start of shift
Test image receptor uniformity	QC technologist	Weekly
Clean cassette-based image receptors	Technologist	Monthly
Compile and review reject analysis data	QA coordinator	Monthly
Verify display calibrations	Clinical engineer	Quarterly
Review QC indicators	QA committee	Quarterly
Verify receptor calibrations	Medical physicist	Semiannual
Verify X-ray generator functions	Medical physicist	Annual

professional (IIP) since this individual plays an integral role in the digital infrastructure of imaging departments. The IIP should be qualified in areas of computer networking infrastructure, databases, data entry, and management systems and have qualifications in a wide range of computer operating systems, internet protocols, and systems related to PACS, RIS, HIS, etc. Furthermore, the ACR, AAPM, and SIIM [18] state categorically that the IIP "have a minimum

of a bachelor's degree in computer science or equivalent. Continuing education and experience in imaging informatics demonstrate that an individual is competent to practice as an imaging informatics professional. Certification through the American Board of Imaging Informatics (ABII) can be used to validate an individual's qualification as a qualified imaging informatics professional" [18].

11.8 Digital Mammography QC

The principles of digital mammography (DM) have been described in Chaps. 7 and 8. The description included equipment for full-field digital mammographic (FFDM) systems such as computed radiography, indirect and direct flat panel detectors, and digital breast tomosynthesis (DBT) which is the focus of Chap. 8. QC for DM and DBT is just as important as QC for the digital radiographic imaging systems described above, and it covers a wide scope of topics. Therefore DM and DBT QC will not be described further in this chapter. In the United States, all mammography equipment are subject to approval by the Food and Drug Administration (FDA). Furthermore, the FDA approved the ACR's "alternative standard request to allow mammography facilities to use the new Digital Mammography Quality Control (QC) Manual and Digital Mammography QC Phantom in routine QC of digital equipment. The new manual and phantom will aid in ensuring uniformity of QC testing" [20]. On July 24, 2017, the ACR implemented the 2016 ACR Digital Mammography Quality Control Manual into the accreditation process, and this new manual is now for accreditation of their 2D full-field digital mammography systems [21].

11.8.1 Parameters for QC Monitoring in Digital Mammography

The ACR provides *recommendations for full-field digital mammography quality control* in which they provide a list of 13 QC tests specifically for the technologist and 17 QC tests to be carried out specifically by the QMP. Details for each test are included under the following headings: Applicability; Minimum Frequency; Performance Criteria; Corrective Action, and Timeframe for Routine QC. For example, for the QC test of phantom image quality, the performance criterion is "must pass image receptor manufacturer's criteria," and the corrective action taken is "immediately" [22]. Table 11.4 provides a task allocation matrix for the technologist for the ACR's recommendations for FFDM QC.

The ACR also lists 18 QC tests that fall within the domain of the QMP. These include tests such as mammography unit assembly evaluation, phantom image quality, missed tissue,

Table 11.4 QC task allocation matrix for technologists for FFDM[a]

Task	Responsibility	Minimum frequency
Daily checklist	Technologist	Daily
Laser printer density consistency	Technologist	Daily (wet), monthly (dry)
Phantom image quality	Technologist	Weekly
Display monitor QC	Technologist	Weekly
Viewboxes and viewing conditions	Technologist	Weekly
Full-field artifacts	Technologist	Monthly
Monthly check lists	Technologist	Monthly
Laser printer artifacts	Technologist	Monthly
Resolution/modulation transfer function (MTF)	Technologist	Quarterly
Repeat analysis	Technologist	Quarterly
Printed image quality	Technologist	Quarterly
Analysis of fixer retention	Technologist	Quarterly
Compression force	Technologist	Semiannually

[a]These tasks are specified by the ACR recommendations for full-field digital mammography quality control [22]

technique chart/AEC evaluation, artifact evaluation, kVp accuracy, beam quality assessment (half value layer), breast entrance exposure and average glandular dose, ghost image evaluation, collimation assessment, resolution/modulation transfer function (MTF), noise, spatial linearity and geometric distortion of the detector, monitor display quality and baseline values, monitor luminance response and viewing conditions, viewbox luminance and room illuminance, laser printer evaluation and baseline values, and finally the evaluation of site's technologist QC [22].

A recent comprehensive position paper on recommendations for digital mammography QA/QC is one by Heggie et al. [23] and published in the *Australasian Physical & Engineering Sciences in Medicine*. The paper offers "the current status of digital mammography QA and recommends test procedures that may be suitable in the Australasian environment" [23].

References

1. Bushong S. Radiologic science for technologists. 11th ed. St Louis, MO: Elsevier; 2017.
2. Katzman GL, Paushter DM. Building a culture of continuous quality improvement in an academic radiology department. J Am Coll Radiol. 2016;13(4):453–60.
3. Seeram E, Brennan P. Radiation protection in diagnostic imaging. Burlington, MA: Jones and Bartlett Learning; 2017.
4. AAPM. Report no. 93 acceptance testing and quality control of photostimulable storage phosphor imaging systems report of AAPM task group 10. College Park, MD: AAPM; 2006.
5. Zarb F, Rainford L, McEntee MF. Image quality assessment tools for optimization of CT images. Radiography. 2010;16(2):147–53.
6. Körner M, Weber CH, Wirth S, Pfeifer KJ, Reiser F, Treitl M. Advances in digital radiography: physical principles and system overview. Radiographics. 2007;27(3):675–86.
7. Seibert JA. Computed radiography/digital radiography: adult. In: Frush DP, Huda W, editors. RSNA categorical course in diagnostic radiology physics: from invisible to visible—the science and practice of x-ray imaging and radiation dose optimization. Chicago, IL: Radiological Society of North America; 2006. p. 57–71.
8. Rowlands JA. The physics of computed radiography. Phys Med Biol. 2002;47(23):R123–6.
9. Willis CE, Weiser JC, Leckie RG, Romlein J, Norton G. Optimization and quality control of computed radiography. PACS design and evaluation. Proc. SPIE. 1994;2164:178–85.
10. Willis CE. Computed radiography, QA/QC. In: Practical digital imaging and PACS. Medical physics monograph no. 28. Madison: Medical Physics; 1999. p. 157–75.
11. Willis CE. Quality assurance: an overview of quality assurance and quality control in the Digital Imaging Department. In: Reiner BI, Seigel EL, Carrino JA, editors. Quality assurance, meeting the challenge in the digital medical enterprise. Great Falls: Society for Computer Applications in Radiology; 2002. p. 1–8.
12. Willis CE, Thompson SK, Shepard SJ. Artifacts and misadventures in digital radiography. Appl Radiol. 2004;33(1):11–20.
13. Bjorn H, Rosanowshy TB, Jnesen C, Wah KHC. Image rejects in general direct digital radiography. Acta Radiol. 2015;4(10):1–6. Open
14. Khafaji MA, Hagi SK. Direct digital radiograph: technicians role in obtaining good images. Saudi Med J. 2014;35(8):879–81.
15. Andersen ER, Jorde J, Yaousse N, Yaqoob SH, Konst B, Seierstad T. Reject analysis in direct digital radiography. Acta Radiol. 2012;53(2):174–8.
16. Jones AK, Heintz P, Geiser W, Goldman L, Jerjian K, Martin M, et al. Ongoing quality control in digital radiography: report of AAPM Imaging Physics Committee Task Group 151. Med Phys. 2015;42(11):6658–70.
17. Seeram E, Davidson R, Bushong S, Swan H. Optimizing the exposure indicator as a dose management strategy in computed radiography. Radiol Technol. 2016;87(4):380–91.
18. ACR-AAPM-SIIM. Technical standard for electronic practice of medical imaging, revised 2017. https://www.acr.org/-/media/ACR/Files/Practice-Parameters/Elec-Practice-MedImag.pdf. Accessed Mar 2018.
19. Willis C. Quality control for digital radiography. In: Seeram E, editor. Digital radiography: an introduction. Clifton Park, NY: Delmar, Cengage Learning; 2011.
20. FDA. FDA approves new ACR digital mammography quality control manual. http://www.aheconline.com/mktg/course_materials/MM_All_Day_Work/i-FDA_Approves_New_ACR_Digital_Mammo_Quality_Control_Manual.pdf. Accessed Mar 2018.
21. ACR. 2017 Digital mammography quality control manual implementation information. https://www.sbionline.org/RESOURCES/WhitePapers/TabId/595/ArtMID/1617/ArticleID/669/Important-2016-ACR-Digital-Mammography-Quality-Control-Manual-Announcement.aspx. Accessed Mar 2018.
22. ACR. The American College of Radiology. Recommendations for full-field digital mammography quality control. https://www.fda.gov/ohrms/dockets/ac/06/briefing/2006-4236b2-05-ACR-Table.pdf. Accessed Mar 2018.
23. Heggie JCP, Barnes P, Cartwright L, Diffey J, Tse J, Herley JMcLean ID, et al. Position paper: recommendations for a digital mammography quality assurance program V4.0. Australas Phys Eng Sci Med. 2017;40(3):491–543.

Dose Optimization in Digital Radiography

12

Abstract

This chapter explored the ICRP's principle of optimization, optimization of the exposure technique factors [kilovolts (kV) and milliampere-seconds (mAs)] used in a digital radiography examination, including the exposure indicator (EI). Secondly, dose optimization tools for image quality assessment were outlined, specifically the method of visual grading of normal anatomy. Finally the chapter provides an example of a research study examining the optimization of the EI as a dose management strategy in digital radiography. The ICRP principle of optimization is intended to protect the patient from unnecessary radiation by using a dose that is as low as reasonably achievable (ALARA). The ultimate goal of optimization is to minimize stochastic effects and to prevent deterministic effects. Three dose optimization approaches were introduced and reviewed. These include optimization of the exposure techniques factor (kV and mAs) and the EI. The literature review on the use of these approaches in computed radiography (CR) found the EI as a dose control mechanism and as a "surrogate for dose management," and the use of the EI has been viewed as an opportunity for dose optimization. Furthermore optimization research has focused mainly on optimizing the kV in CR imaging as a means of implementing the ALARA philosophy.

Generally, these studies have produced "conflicting results." The fourth topic addressed in this chapter is image quality assessment of medical images. These include objective physical measures and observer performance methods such as the use of visual grading of normal anatomy and the various receiver operating characteristic (ROC) methods. One tool that is well established and makes use of the European Guidelines on Quality Criteria for Diagnostic Radiographic Images is visual grading analysis (VGA) procedure. This procedure is used to evaluate image quality based on the visualization and reproduction of defined anatomical structures in images and not on lesion detection. The final section of this chapter examined an empirical research study as an illustration of the use of the EI in the optimization of the radiation dose and image quality for a computed radiography system. Entrance skin doses were measured for phantom models of the pelvis and lumbar spine imaged using the vendor's recommended exposure settings (i.e., the reference doses) as well as doses above and below the vendor's recommended settings for both body parts. Images were assessed using visual grading analysis (VGA). While the dose measurement results revealed strong positive linear relationships between dose and milliampere-seconds (mAs), mAs and inverse exposure indicator (EI), and dose and inverse EI for both body

E. Seeram, *Digital Radiography*, https://doi.org/10.1007/978-981-13-3244-9_12

parts, the VGA showed that optimized values of 16 mAs/EI = 136 for the anteroposterior (AP) pelvis and 32 mAs/EI = 139 for the AP lumbar spine did not compromise image quality. The conclusions drawn of this study suggest selecting that optimized mAs reduced dose by 36% compared with the vendor's recommended mAs (dose) values, and optimizing the mAs and associated EIs can be an effective dose management strategy.

12.1 Introduction

The increasing uses of digital imaging modalities such as computed radiography (CR), flat-panel digital radiography (FPDR), digital fluoroscopy (DF), digital subtraction angiography (DSA), digital mammography (DM), and computed tomography (CT) in medical imaging require a comprehensive understanding of the various technical factors affecting the dose to the patient. There are at least two fundamental issues that are significant for technologists and radiologists using digital imaging systems to be aware of, that is, exposure creep and the wide exposure latitude or dynamic range of the digital detector [1–5]. Dose creep refers to "the risk of increasing patient dose, possibly without being aware of it, or an increase in exposure over time when using digital systems with manual tube settings" [5]. In a similar vein, Cohen et al. [3] explain that dose creep is an "unintended excessive exposure and subsequent unnecessary patient radiation exposure" [3]. The potential harm associated with dose creep is unnecessarily high radiation doses to patients. Furthermore a wide exposure latitude can result in images with high noise levels caused by low exposure or increased radiation doses to patients caused by high exposure [4, 5]. Such understanding is necessary to minimize the risks of radiation biological effects by adhering to the cardinal principles of the radiation protection framework of the International Commission on Radiological Protection (ICRP) [6].

It is important to note that sections of this chapter have been previously published (Radiologic Technology 87(4), 2016) from my PhD thesis, entitled, *Optimization of the Exposure Indicator as a Dose Management Strategy in Computed Radiography.* PhD Dissertation (Charles Sturt University, New South Wales, Australia, 2014).

12.1.1 Biological Effects of Radiation Exposure: An Overview

The biological effects of radiation exposure fall into two categories, namely, stochastic effects and deterministic effects. Stochastic effects are random, and the probability of their occurrence depends on the amount of radiation dose an individual receives. The probability increases as the dose increases, and there is no threshold dose for stochastic effects. Any amount of radiation, no matter how small, has the potential to cause harm. If harm occurs, the damage generally becomes apparent years after exposure; therefore, stochastic effects also are called *late effects*. Examples of stochastic effects include cancer and genetic damage [7–9]. Stochastic effects are considered a risk from exposure to the low levels of radiation used in medical imaging, including CT examinations. Deterministic effects are those for which the severity of the effect (rather than the probability) increases with radiation dose and for which there is a threshold dose. Examples of deterministic effects include skin burns, hair loss, tissue damage, and organ dysfunction [9]. Deterministic effects are also referred to as *early effects* and involve high exposures that are unlikely to occur in medical imaging examinations.

12.1.2 Radiation Protection Philosophy

In medical imaging, the goal is to minimize the probability of stochastic risks and to prevent the occurrence of deterministic effects. To achieve this goal, the ICRP has developed a framework that is guided by three principles. These principles relate to justification, optimization, and dose limitation [10, 11]. While justification refers to the fact that every exposure a patient

receives must have a positive net benefit associated with it, optimization is intended to ensure that all exposures be kept as low as reasonably achievable (ALARA) without compromising the diagnostic quality of the examination. While justification and optimization address individuals who are exposed to radiation, dose limits deal with occupational and environmental exposures and exclude medical exposure. Dose limits are outlined in detail in ICRP Publication entitled *Radiation Protection and Safety in Medicine* [12] and will not be discussed in this chapter.

This purpose of this chapter is to first explain the ICRP's principle of optimization and describe several dose optimization approaches in digital radiography such as optimization of the exposure technique factors [kilovolts (kV) and milliampere-seconds (mAs)] used in a digital radiography examination, including the exposure indicator (EI). Secondly, dose optimization tools for image quality assessment will be outlined, specifically the method of visual grading of normal anatomy. Finally the chapter provides an example of a research study examining the optimization of the EI as a dose management strategy.

12.2 What Is Dose Optimization?

The ICRP principle of optimization is intended to protect the patient from unnecessary radiation by using a dose that is as low as reasonably achievable (ALARA). The ultimate goal of optimization is to minimize stochastic effects and to prevent deterministic effects, as stated earlier. The ICRP recommends approaches to dose optimization associated with radiography equipment and daily operations [12]. Equipment optimization refers to the manufacturer's design and construction of the imaging unit, and operations refer to the options radiologic technologists choose that enable them to work within the ALARA philosophy during the examination [13]. This is the central tenet of ALARA.

Optimization also must address image quality. Specifically, dose optimization strategies must not compromise the image quality required by physicians to diagnose diseases and conditions. Dose optimization is a responsibility of the radiologic technologist's clinical practice. According to Matthews and Brennan [14], the ultimate responsibility for applying optimal technique rests with the individual conducting the examination, and the technologist usually ensures dose optimization. Furthermore, applying ALARA principles to minimize exposure to the patient, along with self and others, is a component of the scope of practice for all medical imaging technologists [14].

When discussing how closely to follow the ALARA principle and balance the need for patient radiation protection with the need for acquiring high-quality diagnostic images, the literature most often uses two terms: reduction and optimization. *Reduction* means to "reduce or diminish in size, amount, extent, or number." *Optimization* means "an act, process, or methodology of marking something (as a design, system, or decision) as fully perfect, functional, or effective as possible" [15]. Optimization therefore is a much more rigorous procedure to ensure the tools and measures used in the process are as objective and effective as possible. In summary, the previously described principle of optimization based on the ICRP radiation protection framework refers to optimization as keeping radiation doses as low as reasonably achievable so as not to compromise the diagnostic quality of the image. Optimization therefore deals with both radiation dose and image quality.

12.3 Dose Optimization Approaches in Digital Radiography

Dose optimization in digital radiography based on image quality requires a thorough understanding of various dose parameters and image quality parameters. As described in Chaps. 3 and 4 and summarized in Chap. 11, the image quality of a digital image has been described in terms of spatial resolution (detail), contrast resolution, noise, detective quantum efficiency (DQE), and artifacts.

12.3.1 Factors Affecting Dose in Digital Radiography: An Overview

Dose parameters on the other hand include several factors affecting the dose to the patient both in digital radiography and in film-screen radiography. These factors include the type of X-ray generator, filtration, collimation, automatic exposure control systems, X-ray beam alignment, the source-to-detector distance (SID), patient thickness, scattered radiation grid characteristics (grid ratio, selectivity, and grid frequency), gonad shielding, sensitivity of the detector, exposure technique factors (kV and mAs), and the repeat rate [2, 9, 16]. It is not within the scope of this chapter to describe all of these factors; however, only the factors under the direct control of the technologist will be reviewed briefly. The major factors under the direct control of the technologist during the examination are the exposure technique factors, the AEC system, collimation, SSD, grid selection, detector sensitivity, gonad shielding, and repeat rate. The following summary about the influence of these factors on dose is noteworthy [2, 9, 17].

1. Increasing mAs and kV increases patient dose proportionally. For example, a twofold increase in mAs increases the dose by a factor of 2. Additionally, doubling the dose requires an increase by the square of the kV change. The algebraic expression for mAs and kV are:
   (a) Dose α mAs.
   (b) Dose α $[kV_{new}/kV_{old}]^2$.
2. With low kVp techniques, more dose is deposited in a larger area of the patient because the photons do not have enough energy to penetrate the entire thickness of the patient. Furthermore, with high kVp techniques, less dose is deposited in the tissues because the photons have a high enough energy to penetrate the tissues and get to the image receptor.
3. AEC systems ensure that a preselected amount of radiation reaches the image receptor and in this manner provide correct exposures.
4. Collimation is intended to protect the patient from unnecessary radiation by limiting the beam field size to the anatomy of interest, thus reducing the volume of tissue irradiated. Restricting the field size to the smallest practicable area results in a decrease in the total radiation energy to the patient, because the surface integral exposure (total radiation) is directly proportional to the size of the irradiated area.
5. Alignment of the radiation field and the light field of the collimator is referred to as beam alignment. If these two fields are not congruent (i.e., in perfect alignment), the anatomical region of interest may not be included because of beam "cutoff," and the patient receives unnecessary radiation because of beam "overlap."
6. Decreasing the SID and hence the source to skin (SSD) will increase the divergence of the beam increases, and this offset has an impact on the concentration of photons on the surface of the patient. A short SID increases the photon concentration per unit area thus increasing the surface exposure to the patient.
7. For scattered radiation grids, as the grid ratio and grid frequency increase, the dose increases. In general, the use of high-frequency grids will result in higher patient doses, because of an increase in the thickness of the lead strips.
8. Detector sensitivity refers to the relative speed of the detector. As detector sensitivity increases, the dose to the patient decreases.
9. Gonad shielding is intended to protect the gonads from primary radiation. These shields must have a thickness of at least 0.5 mm lead equivalent. Furthermore, if gonad shielding is to be effective, the placement of the lead shield on the patient must be accurate to ensure protection of the gonads. For males, it is useful to use the symphysis pubis as the external landmark to guide the placement of the shield. For females, the external landmark 2.5 cm medial to the anterior superior iliac spines can be used for accurate placement of the shield to protect the ovaries from radiation.

10. The repeat rate of an examination increases the dose to the patient. Reject analysis was discussed in Chap. 11.

The following is also noteworthy about image quality characteristics such as noise, spatial resolution, and contrast resolution in relation to the dose [9]:

1. Reducing the noise in an image by a factor of 2 requires an increase in the dose by a factor of 4 (dose is inversely proportional to the noise [2]).
2. Improving the spatial resolution of a digital image (which is affected by the pixel size) by a factor of 2 requires an increase in the dose by a factor of 8 (dose is inversely proportional to the pixel size [3]).
3. With respect to contrast-to-noise resolution in relation to dose parameters, there is "no significant change at different tube voltages in most materials except with high atomic numbers such as iodine (increased at low tube voltage), low-contrast resolution greatly influenced by image noise level" [18].

12.3.2 Optimization of kV

The dose to the patient in a radiographic examination is determined by a host of factors including the primary technical factors of mAs and kV. These factors also play a role in the creation of a diagnostic image. Furthermore the ALARA philosophy requires the correct exposure technique factors to be used for the examination [19]. Additionally, it is significant and important to consider the digital detector response to the X-ray beam energies used in diagnostic radiology (different than film-screen detectors). Moore et al. [16] recommend that compared to film-screen radiography, lower kV should be used in CR imaging [16] since the K-edge of the typical phosphor of a CR imaging plate such as BaFBr/I (barium fluorobromide/iodide) is 37 keV (compared to the K-edge of 50 keV for the phosphor Gd_2O_2S typically used in film-screen radiography). The literature shows "conflicting results" when optimizing the kV and image quality in CR imaging [16, 20].

A review of the literature by Seeram et al. [19] examining studies ranging from as early as 2005 to studies done in 2011 showed that most of the studies focused mainly on chest imaging using CR systems from Fuji, Agfa, Kodak, and Konica-Minolta. These studies have produced "conflicting results" and as such the interested reader should refer to this paper for details of these studies.

An important consideration to pay attention to is the fact that kV affects the penetration of the X-ray beam. Higher kV values must be used to penetrate thicker body parts. In digital radiography, the kV does not affect the image contrast because image post-processing not only affects the density/brightness but also the contrast of the image using the technique of windowing described in Chap. 2. Increasing the kV with the appropriate reduction of the mAs results in less dose to the patient. Since scattered radiation from the patient increases with increasing kV values and will reach the detector, the use of a grid may be needed. Sechopoulos [21] has shown that at 80 kV/675 mA, 120 kV/200 mA, and 140 kV/125 mA, the relative dose to the patient is 104.6, 100, and 87.5, respectively. Finally, the American Association of Radiologic Technologists in their "best practice in digital radiography" White Paper [21] states that with respect to kV, "a best practice in digital imaging is to use the highest kVp within the optimal range for the position and part coupled with the lowest amount of mAs needed to provide an adequate exposure to the image receptor" [22].

12.3.3 Optimization of mAs

The dose to the patient is directly proportional to the mAs used for examinations. Since the mAs also determines the quantity of photons falling on the patient and ultimately affects image noise, therefore optimization of the mAs is also an important consideration in digital imaging. The mAs optimization must consider the assessment of image quality with respect to the dose per image. Increasing the dose per image will decrease the

noise (because of the inverse relationship between mAs and image noise) thus improving image quality [23]. Dosimetric measurements for the AP pelvis and AP lumbar spine are shown in Table 12.1, while Fig. 12.1 shows graphs of the mean dose in mGy plotted as a function of mAs for the AP pelvis and the AP lumbar spine, with respect to reference mAs values (25 mAs for the pelvis and 50 mAs for the pelvis). It is clear that there is a strong positive linear relationship ($r = 0.999$) for both of these body parts [24].

Table 12.1 The results of dosimetric results (mean mGy) for the AP pelvis and the AP lumbar spine measured at several mAs values (data from PhD dissertation [24])

Pelvis		Lumbar spine	
mAs	Mean mGy[a]	mAs	Mean mGy[a]
50	6.45	100	13
40	5.17	80	10
32	4.14	63	7.96
25[b]	3.24	50[b]	6.36
20	2.60	40	5.09
16	2.09	32	4.07
12.5	1.63	25	3.19
8	1.06	20	2.56
6.3	0.83	16	2.05

Abbreviations: *mAs* milliampere seconds, *mGy* milligray
[a]Free-in-air measurements
[b]Reference exposure techniques
All images were obtained at 81 kVp

The wide exposure latitude of digital radiography ranges from about 0.05 mGy to 300 mGy per image [23]; thus optimization should, therefore, assess image quality with respect to the dose per image. A literature review conducted by Seeram et al. [19] cited several studies, such as one that used the lowest mAs value without compromising image quality [25] (due to noise) and highlighted an interesting study by Fauber [20] which showed in detail how the mAs can influence the dose without affecting image quality. Five images were obtained for each of the following exposure groups: 1 mAs, 2 mAs, 4 mAs, 8 mAs (baseline value), 16 mAs, 32 mAs, 64 mAs, and 125 mAs. Fauber [20] explained that "the mAs value to achieve an optimal quality QC phantom image was ultimately decreased by 300% for the lowest mAs value and increased 400% for the highest mAs value." The resolution grid on the image of the phantom was used to assess resolution in terms of the number of line pairs.

The results of this study showed that the line pairs/mm were difficult to discriminate at low exposures compared to the baseline value of 8 mAs, and image mottle (noise) was pronounced at low exposures below the baseline value of 8 mAs. The author concluded that the results of this study simply support the literature on dose optimization in digital radiography, in that diagnostic

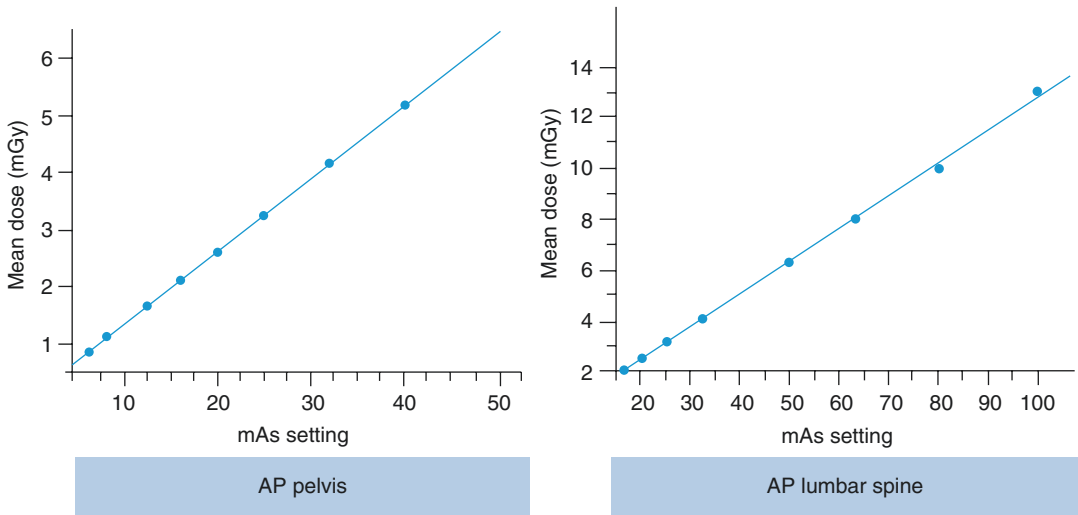

Fig. 12.1 Graphs of the mean dose in mGy plotted as a function of mAs for the AP pelvis and the AP lumbar spine. It is clear that there is a strong positive linear relationship ($r = 0.999$) for both of these body parts (Seeram [24])

quality images can be produced with a wide range of exposures above an optimal level (8 mAs).

12.3.4 Optimization of the Exposure Indicator (EI)

Several different EI schemes were described in Chaps. 3 and 4, and elements of the standardized EI were outlined in Chap. 6. In brief, the EI is a numerical parameter used in digital radiography imaging systems to inform operators whether appropriate radiographic techniques were used for an examination, and therefore it can assist radiologic technologists in the control and management of radiation dose to the patient. The EI is "the key to controlling exposure levels" [4]. In fact, EI can be "used as a surrogate for dose management" [5]. Furthermore, the EI is closely linked with optimizing the kV and the mAs [19]. As noted earlier, the dose to the patient depends on several factors including mAs, kV, collimation, filtration, etc. and does not depend on the EI (not an indicator of patient dose). The EI, however, plays an important role in QC testing (Chap. 11) and indicates whether the correct exposure technique factors are used for the examination.

In terms of using the EI as part of an optimization strategy in digital radiography, the literature is extremely sparse in this area of research. In the previous studies on optimizing kVp and mAs reviewed above, no attempts were made to explore the influence of the kVp and mAs on the EI values. In a literature review by Seeram et al. [19], several studies relating to the optimization of the EI in practice were explored, and the interested reader should refer to this paper for details. The following is noteworthy regarding methods of optimizing the dose to the patient in digital radiographic imaging:

1. The EI has been identified as a dose control mechanism.
2. The EI can be used as a "surrogate for dose management."
3. The use of the EI has been viewed as an opportunity for dose optimization.

Further research is needed to understand the relationship of the EI to exposure techniques and patient exposure [1, 5, 6]. In Sect. 12.6 below, a study on the optimization of the EI as a dose management strategy will be reviewed to provide a small step in using the EI to operate within the ALARA philosophy.

12.4 Dose Optimization Tools for Image Quality Assessment

Image quality goes hand in hand with dose optimization, and the term "dose-image quality optimization" is often used to emphasize that the use of the lowest dose (the optimized dose) must not compromise the quality of the image such that a diagnosis cannot be made. Image quality descriptors such as spatial and contrast resolution, the detective quantum efficiency (DQE), and the noise have been described in Chaps. 3 and 4 and then again summarized in Chap. 11. It is important to understand that a diagnostic image maintains the integrity of these parameters so as not to affect the performance of the observer on basic tasks such as the assessment of the visibility of anatomical structures (the main function of a technologist) as well as lesion detection (the main function of a radiologist) [26, 27].

12.4.1 Methods of Image Quality Assessment

Image quality assessment is a complex task [26, 28] and includes both quantitative objective physical measures coupled with subjective observer performance when evaluating images in a

clinical environment [17]. There are several methods for evaluating image quality in diagnostic imaging based on the "level of ambition" (from lowest to highest). For example, Tingberg and Sjöström [29] explain that while the lowest level would investigate the radiographic technique and measure equipment characteristics and exposure parameters, the highest level would investigate images of patients using a number of different

measurement techniques such as receiver operating characteristics (ROCs) and visual grading analysis. These methods can be placed into two broad categories, namely, *objective physical measures* and *observer performance methods*.

Objective physical measures include measurements that characterize the primary physical characteristics and overall performance of the digital imaging system and relate to the modulation transfer function (MTF), the signal-to-noise ratio (SN), the Wiener spectrum (also known as the noise power spectrum), and the DQE. While these are essentially useful for describing the performance of the imaging system in terms of image quality, they do not relate to all components of the imaging chain [26, 27, 30]. Additionally, in a review of the relationships between physical measurements and user evaluation of image quality, Tapiovaara [31] emphasized that "exact physical measurements cannot supersede subjective evaluation in judging the acceptability of clinical images." These methods are not within the scope of this book and therefore will not be described in this chapter. In medical imaging, the human observer is usually the radiologist; however, the technologist is the first to evaluate the quality of the image, and subsequently the acceptable images are sent to a radiologist for interpretation. It is here where "the information has to be adjusted to conform to the human perception" [30].

Observer performance methods are designed to address whether the observer in viewing and interpreting images is able to discriminate not only the differences between varying states of disease and health but also to describe diagnostically important structures and features, to accurately categorize abnormalities, and to distinguish relevant anatomical structures shown on all images. While these goals are within the function of the radiologist, the last one is also within the function of the technologist. In this regard, there are two categories of observer performance methods: those that examine the function (lesion detection) performed by radiologists and those that are based on the visualization of anatomical structures [26, 29, 30]. While lesion detection makes use of the ROC analysis, visualization of anatomical structures uses visual grading analysis (VGA). The ROC paradigm will not be reviewed here because this chapter focuses on elements relating to the assessment of the visibility of anatomical structures in an image. ROC and its variants (e.g., free-response ROC, localization ROC, alternative free-response ROC) are based on the scientific method and considered the gold standard for image quality assessment. Bath [27], however, points out the following: "conducting these types of studies may be cumbersome, because they are based on the establishment of truth for all cases and normally require a large number of cases in order to produce statistically significant results, which means that the reliability is relatively low. For these reasons, ROC-related methods may not be the method of choice for the local optimization task at a radiation department where the intention is to find the optimal image processing setting or (dose level) for a given examination." Interested readers should refer to the textbook by Chakraborty [32] for a comprehensive description and discussion of these methods.

Visual grading of normal anatomy is a popular, well established, valid, and simple method of subjectively assessing image quality based on the visibility and reproduction of anatomical structures [26, 27, 29, 30, 33]. One of the assumptions of visual grading is that "the visibility of normal anatomy is strongly correlated with the detectability of pathological structures" [30].

Visual grading of normal anatomy will be described briefly in the next section.

12.5 Visual Grading of Normal Anatomy

As pointed out by Bath [27], there are several reasons for using visual grading of normal anatomy in dose optimization studies, and they include not only reliability but also the fact that it is easy to perform compared to ROC studies and does not demand a lot of time on the part of the observer to participate in the assessment of images. Furthermore, visual grading has been used to conduct an image quality audit in clinical practice in

a short timeframe [34]. A major concern of using this approach to an optimization study is that the problem of bias due to the subjectivity involved in visual grading of anatomical images. To address the problem, formal criteria have been developed by a group of expert radiologists and medical physicists and which are now being used in what is referred to as the European Guidelines on Quality Criteria for Diagnostic Radiographic Images [35].

12.5.1 The European Guidelines on Quality Criteria for Diagnostic Images: Overview

An important and significant feature of the European Guidelines on Quality Criteria for Diagnostic Radiographic Images demands that the results of every radiographic examination must be "reproducible and have sufficient sensitivity, specificity, accuracy, and predictive value" [35] and have been adopted for digital radiographic images. In brief, images are defined for specific anatomical landmarks as well as their required level of reproduction in an image to facilitate accurate diagnostic interpretation. For every examination, the criteria are characterized by (1) diagnostic requirements, (2) criteria for radiation dose to the patient, and (3) example of good radiographic technique.

Diagnostic requirements specifically relate to the image criteria, which define the degree of visibility of certain anatomical structures, and four terms are used to explain the meaning of the degree of visibility. These include visualization, reproduction, visually sharp reproduction, and important image details. The meaning of each is shown in Table 12.2. Additionally, the criteria for anatomical structures of the pelvis, for example, that should be seen on an image are listed in Table 12.3. Secondly, the criteria for the dose of the patient require that the weight of a standard-sized adult should be 70 kg and that the absorbed dose for the AP pelvis should be 10 mGy. Finally, the use of manual radiographic techniques is recommended especially in portable radiographic imaging. This notion is supported by Seibert [36] for adult portable chest images who suggested adjusting mAs depending on whether the patient is small, medium, or large. Furthermore, Willis [37] recommended the use of manual radiographic techniques for dose optimization during portable radiography of pediatric patients.

Table 12.2 Definitions of four terms are used to explain the meaning of the degree of visibility, as defined by the European Guidelines on Quality Criteria for Diagnostic Radiographic Images [35]

Term	Definition
Visualization	Characteristic features are detectable but details are not fully reproduced; features are just visible.
Reproduction	Details of anatomical structures are visible but not necessarily clearly defined; detail is emerging.
Visually sharp reproduction	Anatomical details are clearly defined; details are clear.
Important image details	These define the minimum limiting dimensions in the image at which specific or abnormal anatomical details should be recognized.

Table 12.3 Criteria for the anatomical structures for an image of the pelvis (from European Guidelines on Quality Criteria for Diagnostic Radiographic Images [35])

Part	Criteria
AP pelvis	Visually sharp reproduction of the: 1. Greater trochanters 2. Necks of the femora (no foreshortening or rotation) 3. Sacrum and its intervertebral foramina 4. Pubic and ischial rami 5. Sacroiliac joints 6. Cortex/trabecular patterns

12.5.2 Methods of Visual Grading of Anatomical Images

An article by Seeram et al. [38] published in *Radiologic Technology* summarizes the essential features of two methods of visual grading approaches used in the literature. These are the image criteria scoring and visual grading analysis. The image criteria scoring uses the specific image criteria established by the European Guidelines on Quality Criteria for Diagnostic Radiographic Images in visual

grading experiments. Using this approach, the observer's task is to determine whether the image criteria are fulfilled, and subsequently, an image criteria score (ICS) is calculated using the following expression:

$$\mathrm{ICS} = \frac{\sum_{O=1}^{O}\sum_{I=1}^{I}\sum_{C=1}^{C}\mathrm{AR}_{O,I,C}}{O \times I \times C}$$

where $\mathrm{AR}_{O,I,C}$ is the absolute rating for a particular observer (O), image (I), and criterion (C). O, I, and C refer to the number of observers, images, and criteria, respectively [39].

Båth and Månsson [40] point out that "there is no soft transition from 'not fulfilled' to 'fulfilled'; the observer may have difficulties in deciding whether a specific criterion is fulfilled or not when the reproduction of the anatomical structure is close to the decision threshold of the observer. Also, no consideration is taken for how far away from the decision threshold the reproduction is, which may lead to difficulties in interpreting the results." Therefore another method is needed whereby the observer would grade the visibility of anatomical structures using a multistep rating scale. This is the basis of visual grading analysis, and it is intended to improve the "sensitivity to small differences in image quality" [27].

In visual grading analysis, observers evaluate images on the basis of their opinion about the visualization and reproduction of defined anatomical structures using an absolute or a relative rating scale. The absolute scale uses 1, 2, 3, and 4 to rate structure in the image as "not visible," "poorly reproduced," "adequately reproduced," and "very well reproduced," respectively, and the data obtained can be used to calculate a visual grading analysis score (VGASabs) using the following mathematical expression:

$$\mathrm{VGAS}_{\mathrm{abs}} = \frac{\sum_{i=1}^{I}\sum_{s=1}^{S}\sum_{o=1}^{O}G(\mathrm{abs})_{i,s,o}}{O \times I \times C}$$

where G(abs) is the absolute rating for a specific image (i), structure (s), and observer (o). The letters i, s, and o refer to the number of images, structures, and observers, respectively [30].

The relative scale requires that the observer inserts a number in the following statement: "The reproduction of the structure in the image is __the reproduction of the corresponding structure in the reference image." These numbers −2, −1, and 0 to +1 and +2 express the observer's opinion as meaning "much worse than," "worse than," "the same as," "better than," and "much better than," respectively. The data obtained can be used to calculate a relative visual grading analysis score (VGASrel) using the following mathematical expression:

$$\mathrm{VGAS}_{\mathrm{rel}} = \frac{\sum_{o=1}^{O}\sum_{i=1}^{I}\sum_{c=1}^{O}G_{o,i,c}}{O \times I \times C}$$

where $G_{o,i,c}$ is the grading for observer (o), image (i), and criterion (c). The letters o, i, and c refer to the number of observers, the number of images, and the number of criteria, respectively [41, 42].

12.6 Optimization of the EI as a Dose Management Strategy: A Research Study Example

Optimization research studies in medical imaging should meet the following minimum four requirements [43] as shown in the Venn diagram in Fig. 12.2. Four subsets of an optimization research study are important: first the researcher must ensure that the dose to the patient is "safe" and second the level of image quality needed should be determined for the particular diagnostic task. This level deals with the ability of the observer to discriminate between normal and abnormal images [43]. The third important consideration is to acquire images at various exposure levels from high to low and in such a manner so as not to affect the level of image quality required. Finally, reliable and valid methodologies for the dosimetry, image acquisition, and evaluation of image quality should be used. Human

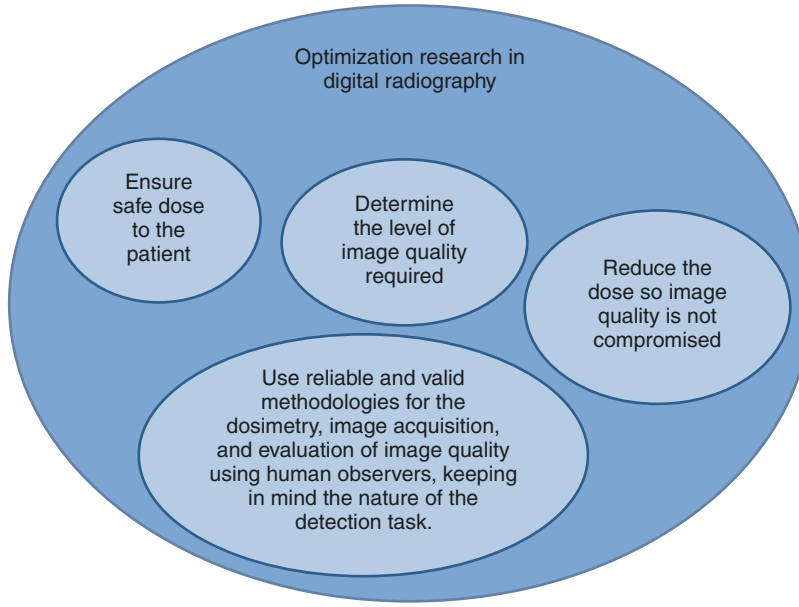

Fig. 12.2 Optimization research studies in medical imaging should meet the following minimum four requirements [43] as shown in the Venn diagram. See text for further explanation

observers should be used for the assessment of image quality, keeping in mind the nature of the detection task. Important considerations in fulfilling these requirements are:

1. The digital imaging system is calibrated to ensure consistent and reliable performance.
2. The dosimeters used to capture dose levels are calibrated.
3. Use phantoms to measure the dose or calculate the absorbed dose to the patient using entrance surface exposures. Established tools and methods of dose measurements should be used.
4. Note the nature of the detection task of the observer. The task of lesion detection (primarily a function of the radiologist) would require a different observer performance method compared to the task of visualization of normal anatomical structures (primarily a function of the technologist).
5. Use the appropriate observer performance method. For example, while the ROC may be considered an effective tool for evaluating observer performance on lesion detection, the VGA may be considered effective for the assessment of visualization of normal anatomical structures. The VGA approach assigns a grade to the image's quality based on comparison with a reference image and is based on the assumption that an observer's ability to see and evaluate normal anatomy correlates to the ability to evaluate pathology or abnormal findings. The ROC approach on the other hand requires observers to determine whether an image contains a lesion or pathology. The observer assigns a grade on a scale (e.g., 1–5) that rates the observer's level of confidence in the decision [44].

An example of an optimization research study that illustrates the requirements defined above is one by Seeram [24] and subsequently published in four papers, one in *Radiography* [19] (International Journal of Radiography and Radiation Therapy), two in *Radiologic Technology* [38] [Journal of the American Society of Radiologic Technologists (ASRT)], and one in the *Journal of Medical Imaging and Radiation Sciences* [45].

12.6.1 The Research Study: An Overview

The title of this research study example is *Optimization of the Exposure Indicator of a CR System as a Radiation Dose Management Strategy* [24]. This study explored the role of the EI in optimizing the dose to the patient in digital radiography using a CR imaging system. Specifically, the purpose of this study was to investigate optimization of the EI of a CR imaging system as a radiation dose management strategy, in keeping with the ALARA principle and to compare the optimized EI with the manufacturer's recommended values for the anteroposterior (AP) pelvis and AP lumbar spine projections.

The methodology involved the use of an anthropomorphic phantom [a transparent pelvis and lumbar spine (L1 to L5) phantom designed to represent an average-sized man approximately 5 ft. 9 in (175 cm) tall and weighing 162 lbs. (73.6 kg) (Radiology Support Devices)]. The phantom contained human skeletal pelvis and lumbar spine parts embedded in anatomically accurate, tissue-equivalent materials with the same radiation absorption characteristics as living tissue. In addition, the methodology involved the exposure technique selection, dose measurement, image acquisition, image quality evaluation, and statistical analysis.

The dosimetry for this study was obtained using a ThinX RAD calibrated dosimeter (Unfors Instruments) and involved measuring the entrance surface dose (ESD) free in air to the phantom for the AP pelvis reflect imaging with 25 mAs and 81 kVp. This is referred to as the *reference exposure technique*. For the AP lumbar spine exposure, technique factors were selected from the manufacturer's technique chart. These factors were listed as 50 mAs and 80 kVp. Again, the kVp setting on the control panel defaulted to 81 kVp. Therefore, 50 mAs and 81 kVp were used as the reference exposure technique for the AP lumbar spine.

For the AP pelvis, 4 ESD measurements were recorded for each of the following mAs values: 6.3, 8, 12.5, 16, 20, 25 (the reference mAs), 32, 40, and 50, all with a fixed kVp of 81. For the AP lumbar spine, 4 measurements were recorded for each of the following mAs values: 16, 20, 25, 32, 40, 50 (the reference mAs), 63, 80, and 100. Thirty-six dose measurements were recorded for the AP pelvis (4 measurements for each of the 9 mAs settings) and 36 for the AP lumbar spine (4 measurements for each of the 9 mAs settings for the lumbar spine). The mean milligray per mAs setting was calculated for each of the nine settings for both the AP pelvis and AP lumbar spine.

The image acquisition involved the following: for the AP pelvis, three images were acquired for each of the following mAs technique settings at 81 kVp: 6.3, 8, 12.5, 16, 20, and 25 mAs (reference mAs). Three images also were acquired using each setting of 81 kVp and mAs of 32, 40, and 50. A total of 27 images were acquired. For the AP lumbar spine, 3 images were then acquired for each of the following mAs technique settings at 81 kVp: 16, 20, 25, 32, 40, and 50 (reference mAs). Three images also were acquired using each setting of 81 kVp and mAs of 63, 80, and 100. A total of 27 images were acquired and processed (Fig. 12.3).

A team of expert observers were used to assess the images based on the visual appearance of noise (quantum mottle) compared to the test images (images recorded with technical factors above and below the vendor's recommended factors). Additionally, dosimetric data were recorded and compared, and the correlation between the dose and EI and dose and mAs was determined using Pearson correlation. Furthermore ANOVA was used to check for statistical significance ($P < 0.05$). Finally Cohen's Kappa analysis was used to check interobserver agreement.

The results of this study showed that observers selected 16 mAs and an EI of 136 for the AP pelvis and 20 mAs and an EI of 220 for the AP lumbar spine as optimized values, compared to the vendor's recommended values.

The VGA procedure showed that although the optimized 16 mA/EI ¼ 136 for the AP pelvis did not compromise image quality, the optimized 20 mA/EI ¼ 220 for the AP lumbar spine compromised image quality. Further analysis of the VGA scores showed that for the AP lumbar spine, it was deemed that 32 mA/EI ¼ 139 was an optimized value where image quality was not

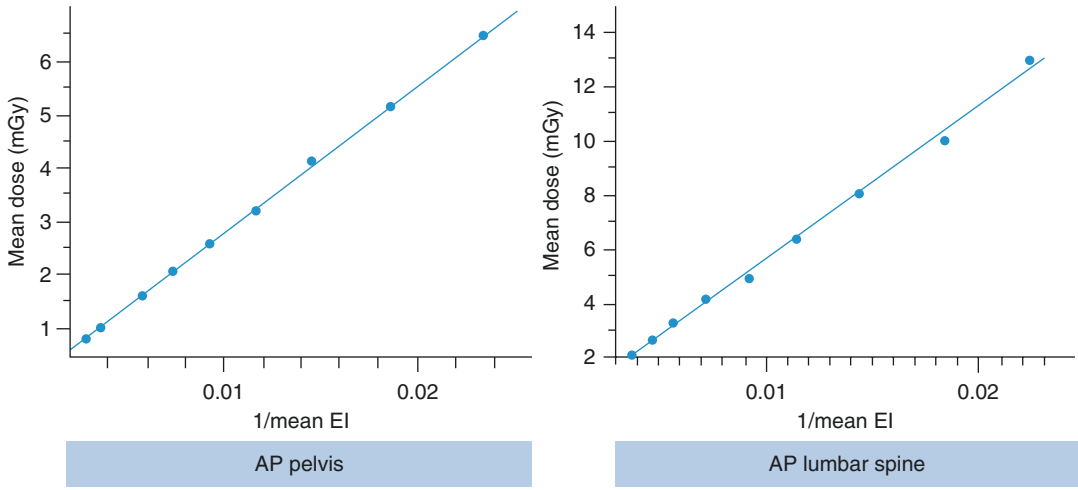

Fig. 12.3 Graphs of the mean dose in mGy plotted as a function of inverse EI for the AP pelvis and the AP lumbar spine. It is clear that there is a strong positive linear relationship ($r = 0.999$) for both of these body parts (Seeram [24])

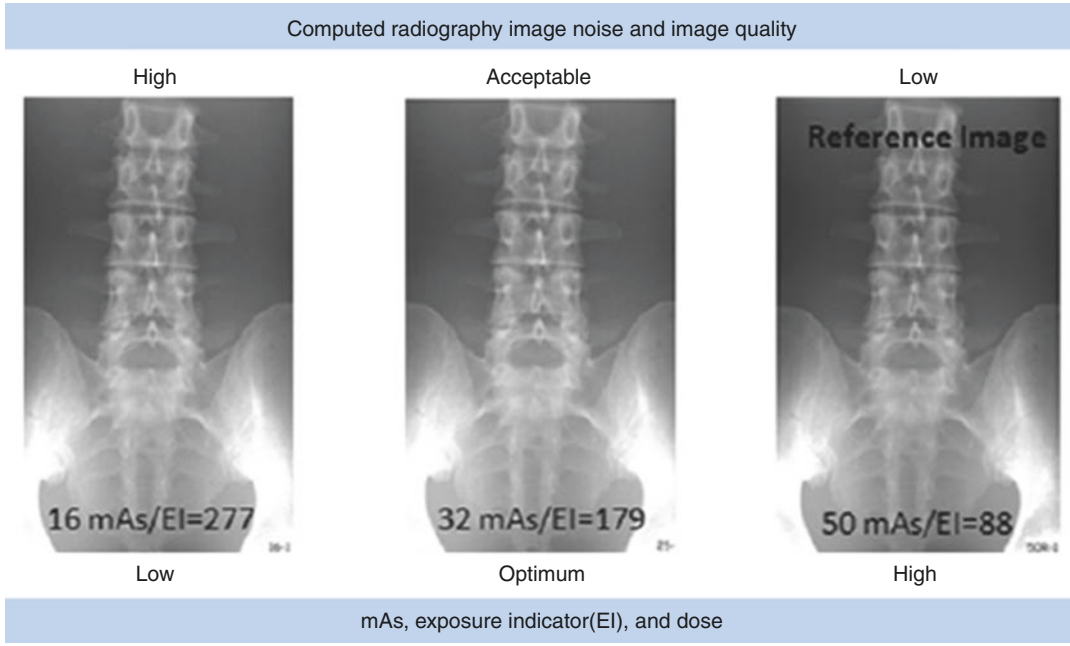

Fig. 12.4 The main findings of the VGA study show that the optimum mA/EI for the AP lumbar spine is 25 mA/179 compared with the manufacturer's reference image obtained at 50 mA and an associated EI value of 88 (Seeram [24])

compromised compared with the vendor's recommended values. The selection of the optimized mA of 16 and 32 for the AP pelvis and AP lumbar spine, respectively, resulted in a dose reduction of 36% compared with the vendor's recommended mA (dose) values without compromising image quality. These images are shown in Figs. 12.4 and 12.5 for the AP lumbar spine and the AP pelvis, respectively.

The conclusions drawn from this research is the mA and its associated EIs can be used as a radiation dose management strategy for implementing the ALARA philosophy in routine daily practice of CR imaging because the

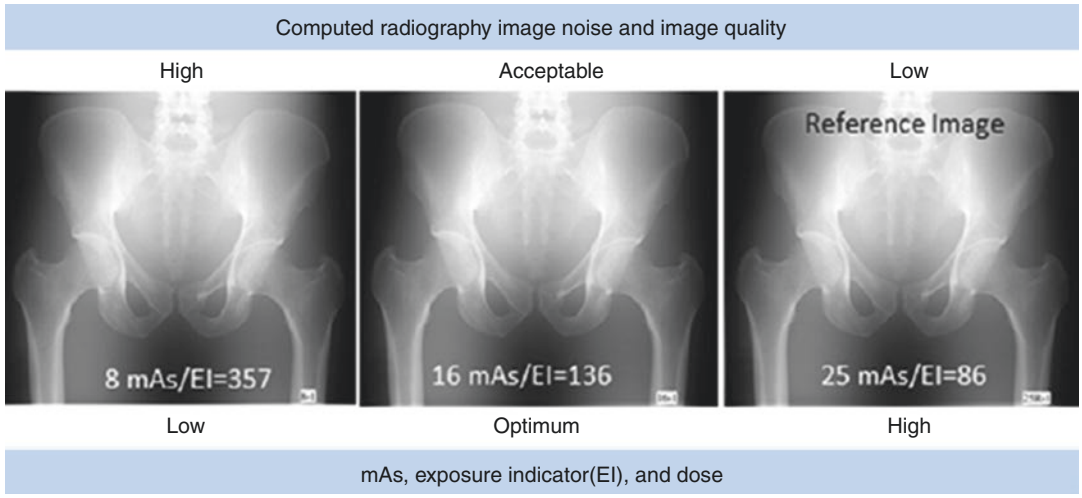

Fig. 12.5 The main findings of the VGA study show that the optimum mA/EI for the AP pelvis is 16 mA/136 compared with the manufacturer's reference image obtained at 25 mA and an associated EI value of 86 (Seeram [24])

dose can be reduced by 36% for both body parts studied in this research compared with the vendor's recommended exposure technique factors.

This study focused on dose optimization using a relatively new digital imaging technology in clinical practice. This topic warrants continual scholarly inquiry as science and technology for digital imaging systems advance. Based on this study, an important future investigation could be performed on the use of a standardized EI. The wide range of EIs and detector exposures used by different manufacturers of digital radiography imaging systems causes confusion among technologists. Therefore, a standardized EI is needed. Today, all digital radiography vendors offer the standardized EI; however, Seibert and Morin noted that "the nuances of this new exposure index standard are now at the beginning of clinical implementation and testing" [2]. Keeping this in mind, the logical next step would be to extend this study to explore how radiologic technologists should implement the standardized EI.

In conclusion, radiologic technologists must understand the distinction between dose reduction and dose optimization in digital radiography as well as the application of the ALARA philosophy and must always be active participants in optimizing patient dose and image quality.

References

1. Gibson DJ, Davidson RA. Exposure creep in computed radiography: a longitudinal study. Acad Radiol. 2012;19(4):458–62.
2. Seibert JA, Morin RL. The standardized exposure index for digital radiography: an opportunity for optimization of radiation dose to the pediatric population. Pediatr Radiol. 2011;41(5):573–81.
3. Cohen MD, Cooper ML, Piersall K, Apgar B. Quality assurance: using the exposure index and the deviation index to monitor radiation exposure for portable chest radiographs in neonates. Pediatr Radiol. 2011;41:592–601.
4. Willis CE. Strategies for dose reduction in ordinary radiographic examinations using CR and DR. Pediatr Radiol. 2004;34(Suppl 3):S196–200.
5. Uffmann M, Schaefer-Prokop C. Digital radiography: the balance between image quality and required radiation dose. Eur J Radiol. 2009;72(2):202–8.
6. International Commission on Radiological Protection (ICRP). Managing patient dose in digital radiology. ICRP publication 93. New York: Elsevier; 2004. p. 1–73.
7. Wolbarst AB, Capasso P, Wyant AR. Medical imaging: essentials for physicians. Hoboken, NJ: Wiley-Blackwell; 2013.
8. Dauer LT, Brooks AL, Hoel DG, Morgan WF, Stram D, Tran P. Review and evaluation of updated research on health effects associated with low-dose ionizing radiation. Radiat Prot Dosimetry. 2010;140(2):103–36.
9. Bushong S. Radiologic science for technologists: physics, biology and protection. 11th ed. St Louis, MO: Mosby-Elsevier; 2017.
10. The 2007 Recommendations of the International Commission on Radiological Protection. ICRP publication 103. Ann ICRP. 2007;37(2-4):1–332.

11. ICRP publication 105. Radiological protection in medicine. Ann ICRP. 2007;37(6):1–63.
12. Radiological protection and safety in medicine. Recommendations of the International Commission on Radiological Protection. Publication no. 103. Ann ICRP. 2007;37:2–4.
13. Hricak H, Brenner DJ, Adelstein SJ, Frush DP, et al. Managing radiation use in medical imaging: a multifaceted challenge. Radiology. 2011;258(3):889–905.
14. Matthews K, Brennan P. Optimization of x-ray examinations: general principles and an Irish perspective. Radiography. 2009;15:262–8.
15. Reduction, optimization. Merriam-Webster dictionary. Online english language dictionary. www.merriam-webster.com/dictionary. Accessed Mar 2018.
16. Moore CS, Saunderson JR, Beavis AW. Investigating the exposure class of a computed radiography system for optimisation of physical image quality for chest radiography. Br J Radiol. 2008;82(981):771–7.
17. Bushberg JT, Seibert JA, Leidholdt EM, Boone JM. The essential physics of medical imaging. 3rd ed. Philadelphia, PA: Wolters Kluwer/Lippincott Williams & Wilkins; 2012.
18. Goo HW. CT radiation dose optimization and estimation: an update for radiologists. Korean J Radiol. 2012;13(1):1–11.
19. Seeram E, Davidson R, Bushong S, Swan H. Radiation dose optimization research: exposure technique approaches in CR imaging: a literature review. Radiography. 2013;19(4):331–8.
20. Fauber TL, Cohen TF, Dempsey MC. High kilovoltage digital exposure techniques and patient dosimetry. Radiol Technol. 2011;82(6):501–9.
21. Sechopoulos I. In: Shaw CC, editor. Cone beam CT. Boca Raton, FL: CRC Press; 2014.
22. Herrmann TL, Fauber TL, Gill J, Hoffman C, Orth DK, Peterson PA, Prouty RR, Woodward AP, Odle TG. Best practices in digital radiography: white paper, ASRT. Albuquerque, NM: American Society of Radiologic Technologists. p. 2012.
23. Marshall NW. Optimisation of dose per image in digital imaging. Radiat Prot Dosim. 2001;94(1e2):83–7.
24. Seeram E. Optimization of the exposure indicator as a dose management strategy in computed radiography. PhD dissertation, NSW, Australia, 2014.
25. Matthews K, Brennan PC. Optimisation of x-ray examinations: general principles and an Irish perspective. Radiography. 2009;15:262–8.
26. Månsson LG. Methods for the evaluation of image quality: a review. Radiat Prot Dosim. 2000;90(1-2):89–99.
27. Båth M. Evaluating imaging systems: practical applications. Radiat Prot Dosim. 2010;139(1-3):26–36.
28. Martin CJ. Radiation dosimetry for diagnostic medical exposures. Radiat Prot Dosim. 2008;128(4):389–412.
29. Tingberg A, Sjöström D. Optimisation of image plate radiography with respect to tube voltage. Radiat Prot Dosim. 2005;114(1-3):286–93.
30. Sund P, Båth M, Kheddache S, Månsson LG. Comparison of visual grading analysis and determination of detective quantum efficiency for evaluating system performance in digital chest radiography. Euro Radiol. 2004;14(1):48–58.
31. Tapiovaara MJ. Review of relationships between physical measurements and user evaluation of image quality. Radiat Prot Dosim. 2008;129(1-3):244–8.
32. Chakraborty DP, editor. Observer performance methods for diagnostic imaging: foundations, modelling, and applications with R-based examples. Boca Raton: CRC Press; 2018.
33. Smedby O, Fredrikson M. Visual grading regression: analysing data from visual grading experiments with regression models. Br J Radiol. 2010;83(993):767–75.
34. Tesselaar E, Dahlström N, Sandborg M. Clinical audit of image quality in radiology using visual grading characteristics analysis. Radiat Prot Dosim. 2016;169(1-4):340–6.
35. Commission of the European Communities. European guidelines on quality criteria for diagnostic radiographic images. Brussels: European Commission; 1996.
36. Seibert JA. Digital radiography: image quality and radiation dose. Health Phys. 2008;95(5):586–98.
37. Willis CE. Optimizing digital radiography of children. Eur J Radiol. 2009;72(2):266–73.
38. Seeram E, Davidson R, Bushong S, Swan H. Optimization of the exposure indicator as a dose management strategy in computed radiography. Radiol Technol. 2016;87(4):380–91.
39. Lanhede B, Båth M, Kheddache S. The influence of different technique factors on image quality of chest radiographs as evaluated by modified CEC image quality criteria. Br J Radiol. 2002;75(889):38–49.
40. Båth M, Månsson LG. Visual grading characteristic (VGC) analysis: a non-parametric rank invariant statistical method for image quality evaluation. Br J Radiol. 2007;80(951):169–76.
41. Geijer H, Persliden J. Varied tube potential with constant effective dose at lumbar spine radiography using a flat-panel digital detector. Radiat Prot Dosim. 2005;114(1-3):240–5.
42. Tingberg AM. Quantifying the quality of medical X-ray images: an evaluation based on normal anatomy for lumbar spine and chest radiography [dissertation]. Lund: Lund University, Department of Radiation Physics; 2000.
43. Mattsson E. Optimisation strategies in medical x-ray imaging. Radiat Prot Dosim. 2005;114:1–3.
44. Tingberg A, Båth M, Håkansson M. Evaluation of image quality of lumbar spine images: a comparison between FFE and VGA. Radiat Protect Dosimetry. 2005;114(1-3):53–61.
45. Seeram E. The new exposure indicator for digital radiography. J Med Imag Rad Sci. 2014;45:144–58.