MEDICAL IMAGING

Principles, Detectors, and Electronics

Edited by

Krzysztof Iniewski

Redlen Technologies Inc.

WILEY

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Published by John Wiley & Sons, Inc., Hoboken, New Jersey

Published simultaneously in Canada

Library of Congress Cataloging-in-Publication Data:

Iniewski, Krzysztof
Medical imaging electronics/Krzysztof Iniewski.
p. cm.
Includes bibliographical references and index.
ISBN 978-0-470-39164-8 (cloth)
1. Imaging systems in medicine. 2. Medical electronics. I. Title.
R857.O6I55 2009
616.07′54—dc22

2008045490

Printed in the United States of America

10 9 8 7 6 5 4 3 2 1

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PREFACE

The ability to peer into the human body is an essential diagnostic tool in medicine and is one of the key issues in health care. Our population is aging globally; for example, over 20% of the population in Japan is already over 65 years old. Older people require many more imaging investigations than do younger ones. Cancer and heart disease are the number-one killers, approaching 40% of all deaths. Improved image quality becomes essential for effective diagnostics in these cases. Shorter examination times, shift to outpatient testing, and noninvasive imaging are rapidly needed. The challenges to contain health-care costs are enormous, and technology solutions are needed to address them.

This book addresses the state-of-the-art in hardware design in the context of medical imaging of the human body. There are new exciting opportunities in ultrasound, magnetic resonance imaging (MRI), X-ray, computed tomography (CT), and nuclear medicine (PET/SPECT). Emerging detector technologies, circuit design techniques, new materials, and innovative system approaches are explored. This book is a must for anyone serious about electronics in a health-care sector.

There are four major imaging modalities described in this book. Their effective signal positions on the electromagnetic spectrum vary from kilohertz (kHz) for ultrasound, through gigahertz (GHz) for magnetic resonance imaging (MRI), to 10^{18} Hz for X-ray/computed tomography (CT) and nuclear medicine, over 15 orders of magnitude variation! Despite their vastly different frequencies and principles of operation, there are numerous commonalities in signal processing of signals received by these imaging detectors, such as signal amplification, filtering, multiplexing, and analog-to-digital conversion. These hardware commonalities among imaging techniques merit putting all related knowledge and know-how into one publication. I sincerely hope that this book will help improve the understanding of medical imaging electronics and stimulate further interest in the development and use of this equipment to benefit us all.

KRZYSZTOF (KRIS) INIEWSKI

Vancouver 2008

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Krzysztof (Kris) Iniewski is managing R&D chip development at Redlen Technologies Inc., a start-up company in British Columbia. His research interests are in VLSI circuits for medical and security applications. From 2004 to 2006 he was an Associate Professor in the Electrical Engineering and Computer Engineering Department of University of Alberta where he conducted research on low power wireless circuits and systems. During his tenure in Edmonton, he wrote a book for CRC Press, entitled *Wireless Technologies: Circuits, Systems, and Devices*.

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PART I
X-Ray Imaging and Computed Tomography

1 X-Ray and Computed Tomography Imaging Principles

KRZYSZTOF INIEWSKI

1.1. INTRODUCTION TO X-RAY IMAGING

X-ray imaging is a well-known imaging modality that has been used for over 100 years since Röntgen discovered X-rays based on his observations of fluorescence. His initial results were published in 1885, and reports of diagnoses of identified fractures shortly followed. A year later, equipment manufacturers started selling X-ray equipment. Today, X-ray and its three-dimensional (3D) extension, computed tomography (CT), are used commonly in medical diagnosis.

X-rays are high-energy photons. Their generation creates incoherent beams that experience insignificant scatter when passing through various media. As a result, X-ray imaging is based on through transmission and analysis of the resulting X-ray absorption data. Typically, X-rays are detected through a combination of a phosphor screen and a light-sensitive film, as shown in Fig. 1.1. The current system, which has been used for mammography and radiography for many years, provides a good-quality analog image that is not compatible with digital storage and transmission requirements of the modern digital era. A slight variation of this common technique is used in fluoroscopy where image intensifier is used as transition stage to supply signals to CMOS cameras producing an analog image directly on a TV screen. Multiple conversions steps in this case from X-rays to electrons to light to camera display lead to poor image quality.

An alternative to the conventional detection technique, also shown in Fig. 1.1, uses a digital detector that converts X-ray photons directly into an electrical signal of digital nature. Chapter 2 in this book discusses an example of this direct detection technology using a large-area active matrix flat panel based on the amorphous silicon (a-Si). Having a digital image leads to lower storage cost and ease of electronic transmission in a future e-Healthcare era.

Medical Imaging: Principles, Detectors, and Electronics, edited by Krzysztof Iniewski

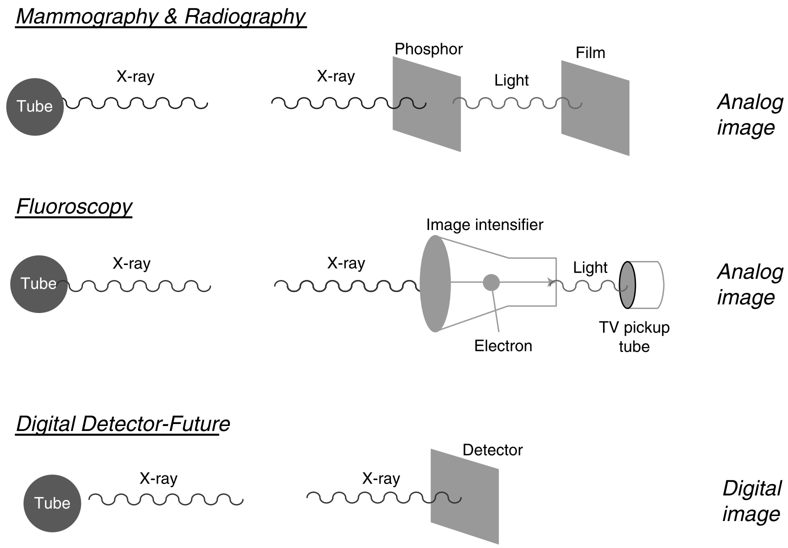

Fig. 1.1. Typical X-ray detection methods used today that provide an analog image (**top** and **middle**) and that in the future will use a digital detector (**bottom**). (From http://www.ecse.rpi.edu/censsis/.)

There are some important differences in characterization of film-based imaging and digital imaging. In this chapter and the rest of the book we will focus on digital imaging because it is a more modern technique which with time is expected to completely replace film-based imaging the same way that digital cameras have displaced analog films in consumer cameras. In digital imaging we use terms such as brightness, dynamic range, linearity, or signal-to-noise ratio instead of density, latitude, film speed, or image sharpness, the terms associated with film-based technology.

Digital X-ray detectors can operate in two regimes: photon counting and integration. In the photon counting mode, each individual photon is detected; and if its energy is higher than the set threshold, the photon is counted with its corresponding energy registered if desired. In the integrating mode the charge generated by the incoming photon is integrated in a selected time interval. Due to this principle of operation, a count rate in the photon counting mode is limited, typically to 10^6 counts per second (c/s), while it is virtually unlimited for the integrating mode. The photon counting mode can detect smaller signals, down to an individual photon, and offers a higher dynamic range (typically 10^6) compared to the integrating mode (10^4). Advantages of the photon counting mode include higher detector quantum efficiency (DQE), lower electronic noise, no need for signal digitization, and possibility of energy discrimination. The integrating mode, in turn, can operate with high count rates, and it is simple and inexpensive to implement. Chapter 3 in this book discusses differences in both photon counting and integration modes of operation in more detail.

While operation of modern X-ray based scanners can be quite complex, a basic principle behind X-ray imaging is quite simple. The technique relies on analyzing attenuation data of the object (patient) that undergoes X-ray exposure. Because different materials (internal organs) experience different levels of X-ray intensity attenuation, an image corresponding to these properties can be readily created. The attenuation characteristics are governed by the so-called Beer–Lambert law, which is expressed as follows:

$$I(z) = I_0 * \exp(-\mu z) \tag{1.1}$$

where $I(z)$ is the X-ray intensity at the detector, I_0 is the X-ray intensity at the source, z is the distance between the source and the detector planes, and μ is the attenuation coefficient that has a different value for different materials. By measuring $I(z)$ for a set of detectors, one can establish the corresponding value of the attenuation coefficients that give a representation of the image. X-ray imaging is particularly good for providing a contrast between soft and hard tissues, because the attenuation coefficient has a quite different value in both media; hence one of the first applications was to identify fractured bones.

To operate as a diagnostic technique, X-ray imaging needs a radiation source, a means of interactions between the X-ray beam and the object to be imaged, ways of registration of the radiation carrying information about the object, and finally the ability to convert that information into an electrical signal. Although widely used and inexpensive, standard X-ray technique have quite severe limitations. First, 3D structures are collapsed into 2D images, leading to highly reduced image contrast. Second, it is difficult to image soft tissues due to small differences in attenuation coefficients. Finally, standard film-based technology does not provide quantitative data and requires specialized training for accurate image assessment.

Fortunately, a 3D extension of 2D X-ray technology, called computed tomography (CT), was invented in 1972 and is in widespread use today. A basic principle behind CT is to take a large number of X-ray images at multiple angles and, based on that information, calculate the 3D image of the imaged object. CT hardware used for this application is typically called a CT scanner and is similar to an ordinary X-ray machine, albeit with much more computational power. With today's multiple-row detector helical CT scanners, 3D images can be obtained with spatial resolution approaching that of conventional radiographic images in all three dimensions.

This chapter is organized as follows. The radiation source, a well-known X-ray tube, is discussed briefly in Section 1.2. Details of interaction between photons and the object, which include absorption, reflection, scattering, and diffraction, are considered in Section 1.3. Detectors used to register the radiation events are discussed in Section 1.4, while conversion of electrical signals is mentioned in Section 1.5. Principles of computed tomography (CT) are introduced in Section 1.6, while CT scanner design is described in Section 1.7. Extension of X-ray imaging that takes into account photon energy, referred to as "color" X-ray imaging, is discussed in Section 1.8 followed by summary of future trends in Section 1.9. For more details on X-ray imaging modalities, the reader is referred to numerous books on this subject [1–7].

1.2. X-RAY GENERATION

A typical X-ray tube is shown in Fig. 1.2. Generation of X-rays depends on thermionic emission and acceleration of electrons from a heater filament. During that process, electrons emitted from cathode are accelerated by anode voltage. Kinetic energy loss at an anode is converted to X-rays. The relative position of an electron with respect to the nucleus determines the frequency and energy of the emitted X-ray.

X-rays produced in an X-ray tube contain two types of radiation: *Bremsstrahlung* and characteristic radiation. The word *Bremsstrahlung* is retained from the German language to describe the radiation that is emitted when electrons are decelerated. It is characterized by a continuous distribution of X-ray intensity and shifts toward higher frequencies when the energy of the bombarding electrons is increased. Characteristic X-rays, on the other hand, produce peaks of intensity at particular photon energies as shown in Fig. 1.3. In practice, emitted radiation is filtered, intentionally or not, producing high-pass filter response as low-energy radiation is completely attenuated. As a result, the final X-ray spectrum has band-pass type characteristics with several local peaks superimposed on it (Fig. 1.4).

The filtering effect shown in Fig. 1.4 is intentional, used to cut off X-ray energies below 20 keV in the shown example. A similar effect can be achieved unintentionally if the gap between the source and the detector is large. Figure 1.5 shows transmission characteristics through air. While 40-keV radiation is not affected by the air gap, 10-keV rays are severely attenuated, and the degree of their attenuation is dependent on the distance.

X-ray generation is a fairly inefficient process because most of the electrical power ends up as heat at the anode. Therefore, an X-ray tube is also a heater, and heat

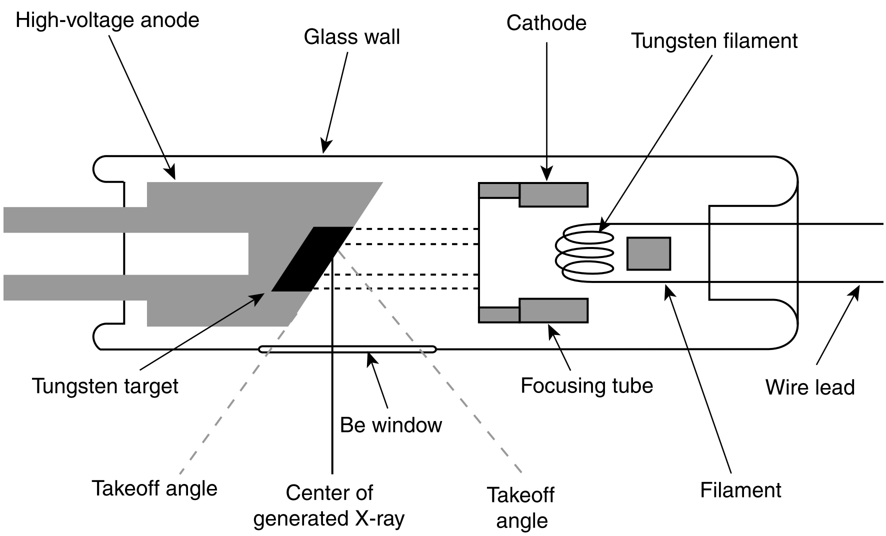

Fig. 1.2. X-ray tube. From http://www.siint.com/en/technology/xrf_descriptions1_e.html.

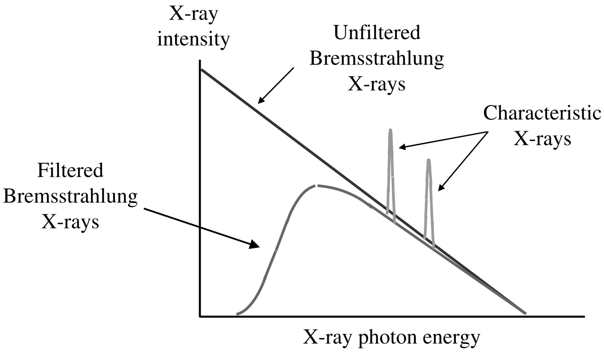

Fig. 1.3. Schematic representation of X-ray intensity frequency characteristics.

extraction problems are primary problems in the equipment design and manufacturing. In addition, only a few percent of the generated X-rays end up being absorbed at the detector because the X-ray beam is not collimated and photons are radiated in all possible directions. X-ray photon energy is related to acceleration voltage; so if the acceleration voltage is 20 kV, it will produce 20-keV photons. Clearly, X-ray-based equipment is clearly not suitable for home use! The total number of photons generated is proportional to the cathode current, which typically is several milliamperes. A typical X-ray system uses step-up transformers to produce high-voltage (HV) as schematically shown in Fig. 1.6.

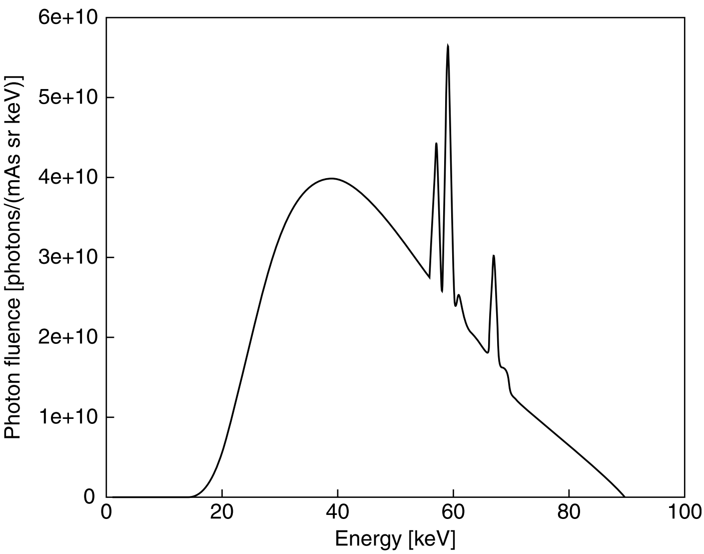

Fig. 1.4. Simulated X-ray intensity characteristics for a 90-keV tube with a 1.5-mm Be and 2.7-mm Al filter. (From Roessl and Proksa [8], with permission.)

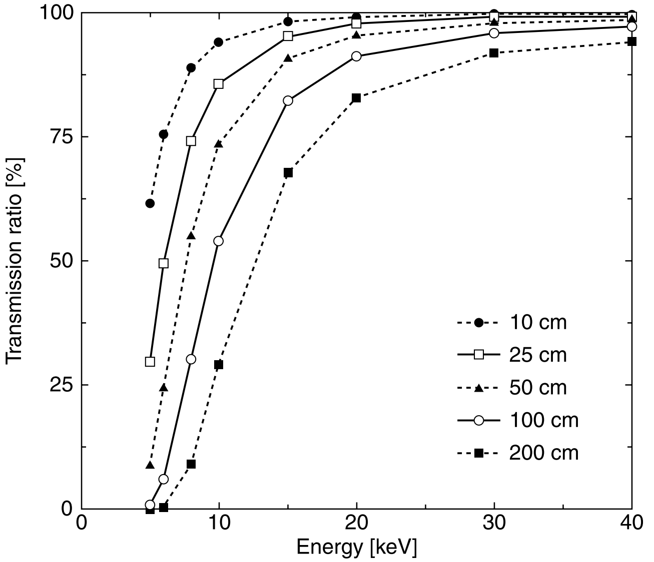

Fig. 1.5. X-rays transmission characteristics for the air path as a function of X-ray energy. (From Miyajima and Imagawa [9], with permission.)

X-ray tubes used in computed tomography (CT) are subjected to higher thermal loads in than in any other diagnostic X-ray application. In early CT scanners, stationary anode X-ray tubes were used, since the long scan times meant that the instantaneous power level was low. Long scan times also allowed significant heat dissipation. Shorter scan times in later versions of CT scanners required high-power

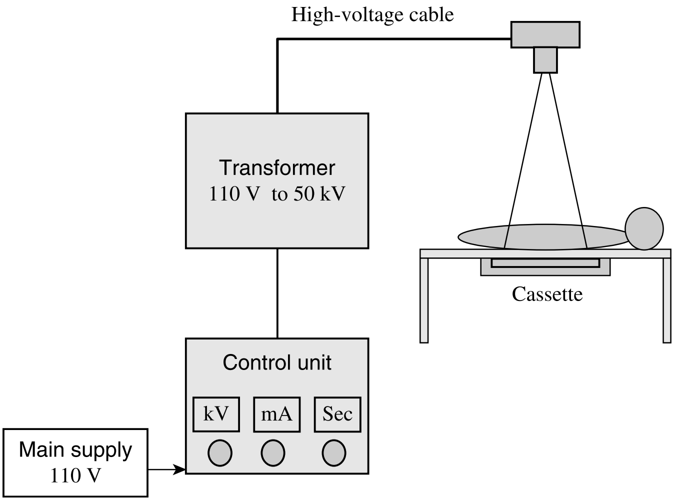

Fig. 1.6. Schematic representation of a standard X-ray system.

X-ray tubes and use of liquid-cooled rotating anodes for efficient thermal dissipation. The recent introduction of helical CT with continuous scanner rotation placed even more demands on X-ray tubes; this is clearly a challenging engineering problem because the dissipated power is in the kilowatt range.

X-rays represent ionizing radiation that at significant dose will cause tissue damage. The traditional unit of absorbed dose is the rad. 1 rad is defined as the amount of X-ray radiation that imparts 100 ergs of energy per gram of tissue or, as re-stated in SI units, causes 0.01 joule of energy to be absorbed per kilogram of matter. As a frame of reference, a typical chest X-ray exposure is about 50 mrads, while exposure of 50 rads causes radiation sickness. In the SI system, rad is now superseded by gray, with the following simple relationship between the two: 1 gray equals 0.01 rad.

1.3. X-RAY INTERACTION WITH MATTER

X-rays interact with matter in several ways that can be divided into absorption and scattering effects. Primary effects at energies of interest in medical applications are photoelectric effect, Compton scatter, and coherent scatter. In the photoelectric effect, the energy of an X-ray photon is absorbed by an orbital electron, which in turn is ejected from an atom. During this process, X-rays are converted into electric charges, a process very useful for radiation detection. Scattering effects can be of Compton nature, where some energy loss is involved, and coherent, without any energy loss. In Compton scatter, some of the X-ray energy is transferred to an electron, and the X-ray photon travels on with an altered direction and less energy. The Compton process might sometime be utilized in medical imaging in so-called Compton cameras, but frequently it is an undesired effect. As opposed to the Compton effect in coherent scatter, all X-ray energy interacts with the atom, but is later re-radiated with same energy in an arbitrary direction. As a result, the photon changes direction but still carriers the same energy, a process quite detrimental to medical imaging because the original path of photon from a source to a detector is altered.

The relative probability of above processes is dependent on the photon energy and characteristics of the matter with which it interacts. In order to focus our discussion here, we will discuss some details of photon interaction with a semiconductor material called CZT. CZT stands for cadium zinc telluride and is currently considered as the most promising detector material for X-ray and γ-ray direct detection in medical imaging for reasons that are explained later in this chapter. The relative probability of absorption/effect is plotted in Fig. 1.7. The photoelectric effect is a dominant one in the considered energy range of 20–300 keV; although at higher energies, Compton scattering becomes equally probable. At the energy of 122 keV, which represents a characteristic cobalt radiation line, the photoelectric effect has 82% probability of happening, Rayleigh scattering 7%, and Compton scattering 11%.

Note that the photoelectric line shows an interesting behavior in the 20- to 40-keV range due atomic structure. The corresponding attenuation length, shown in Fig. 1.8, varies from 0.05 to 0.17 mm. This indicates that even a thin CZT detector will effectively absorb all radiation in that energy range.

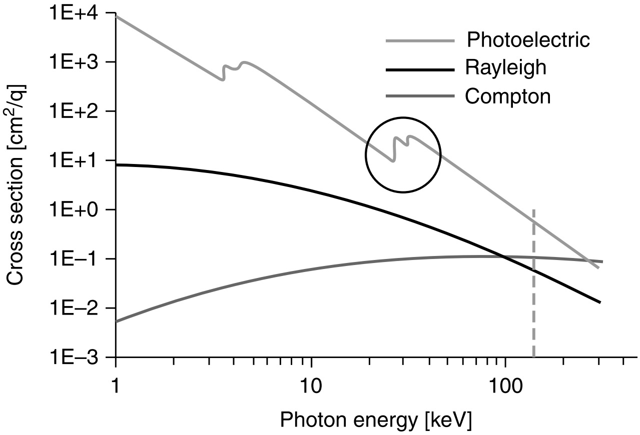

Fig. 1.7. Effective photon cross sections for the photoelectric effect, Rayleigh scattering, and Compton scattering in CZT. The dashed vertical line indicates the 122-keV cobalt line.

The photoelectric effect is one of the energy loss processes where the photon effectively disappears after the interaction. A complete absorption of the photon energy is the desired effect for X-ray detection. The name photoelectron comes from a process of ejecting an electron from one of the atomic shells of the media. After the ejection of the photoelectron, the atom is ionized. The vacancy in the bound shell is refilled with

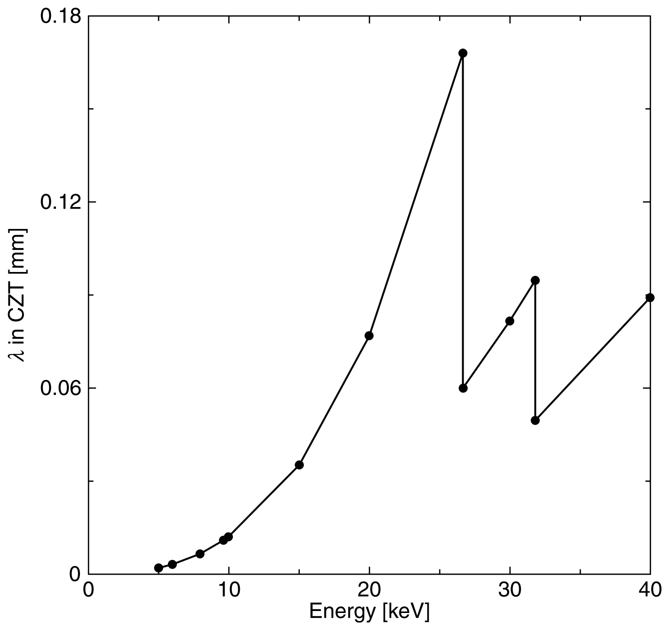

Fig. 1.8. Attenuation length in CZT for X-ray energies from 5 to 40 keV.