

MEDICAL AND INDUSTRIAL TOMOGRAPHY

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Summary

This chapter describes the principles and practice of tomographic imaging and some of its medical and industrial applications. Conventional X-radiographs suffer from interference between overlapping shadows of features situated at various depths within the sample being studied since such images contain almost no depth information. However by recording multiple images from many different viewpoints around an object the problem of neighboring planes was solved with the practical realization of computer tomography in the 1970s. This technique had a revolutionary impact on soft tissue radiography in medicine and inspired tomographic imaging by other means, notably Magnetic Resonance Imaging, as well as spreading into non-medical areas of application. The theoretical basis of reconstructive tomography is briefly outlined and some details of its practical implementation are given. Finally, medical and industrial

applications of this technology are illustrated by a variety of examples.

1. Introduction

In November 1896, at the University of Würzburg, Wilhelm Conrad Röntgen made the accidental discovery of a new form of invisible radiation, while experimenting with electrical discharges in low-pressure gases. He called these X-rays in view of their initially mysterious properties and they continue to be described this way throughout most of the world, although they are named after their discoverer in German speaking countries. They traveled in straight lines through matter and could be recorded photographically. But their most remarkable property was their ability to penetrate substantial thicknesses of optically opaque materials such as biological tissues, and the famous X-ray shadow-graph of Bertha Röntgen's hand, Figure 1, opened up the new field of radiography.



Figure 1: Röntgen recorded this radiograph of the ring hand of his wife Bertha within a few days of his discovery of X-rays in December 1895.

The great contrast between bone and soft tissue suggested the immediate application to medical diagnosis of broken limbs, and later to other parts of the body. However, its great strength in visualizing dense bony structures was also a drawback when attempting to examine low-density soft tissues such as heart, lung and other organs, which are obscured by the strong shadows of the ribs and spine, Figure 2; study of the brain through the protective skull is also greatly handicapped.

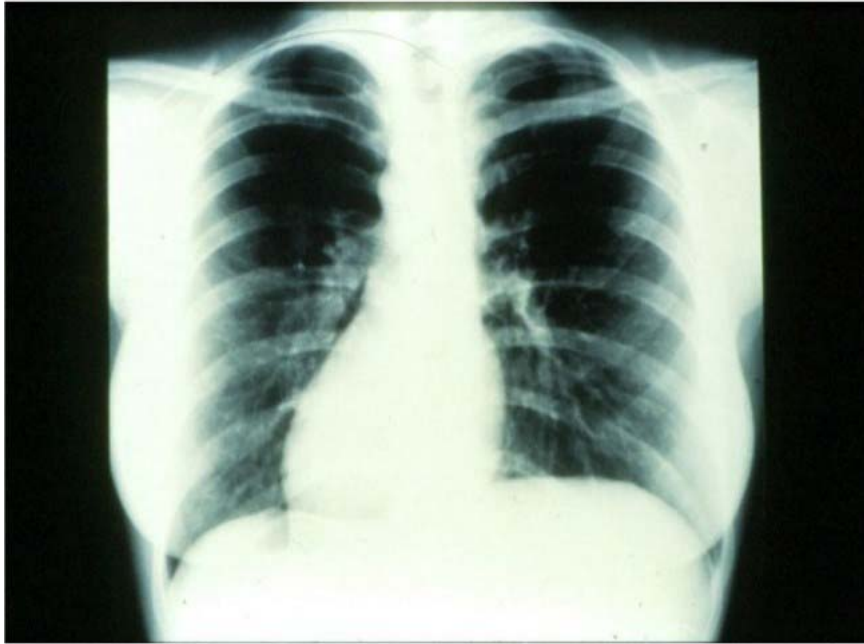


Figure 2: Conventional chest radiograph showing how strong X-ray absorption by the rib-cage obscures details of underlying soft tissues.

This difficulty was partially resolved in 1921 with the invention of focal-plane tomography (Greek *tome* slice, *graphie* writing) by Bocage. In this technique the X-ray source and detector are moved in opposite directions on either side of the patient such that the image of a selected plane of the body remains stationary on a photographic film while contributions from all other planes are blurred out by the relative motion. Interference due to shadowing by overlying tissues is reduced by this process but image contrast is still degraded to some extent. This problem was overcome in 1968 with the invention of the first computerized tomographic (CT) scanner by Hounsfield; this was conceived independently of the work by Cormack who had demonstrated the principles of reconstructive tomography with a simple laboratory rig several years earlier. Their joint contribution was recognized by the award of the Nobel Prize for Medicine in 1979. Following the early commercial development of CT scanners, it was discovered that the basic elements of reconstructive tomography with X-rays had been published as early as 1956 by Tetel'baum whose work, based on analog electronics, proved to be impractical and lay unrecognized in the Soviet technical literature for many years. It required the development of the mini-computer in the 1960s to render practicable the digital image processing approach of Cormack and Hounsfield.

Hounsfield's original head scanner immediately obviated the traumatic invasive techniques hitherto used for radiography of the brain, and its ability to image subtle changes in density revolutionized soft tissue radiography. The instant success of tomographic imaging with X-rays greatly stimulated developments with other modalities (e.g. gamma-rays, ultrasound, MRI) and these have now taken their place alongside X-ray CT as complementary radiographic techniques. The physical and mathematical bases of reconstructive tomography are briefly considered in the following section before various practical implementations and applications of these principles are described. The majority of this review deals with tomographic imaging

based on ionizing radiation but some brief references to non-ionizing radiation techniques are also included.

2. Principles of Reconstructive Tomographic Imaging

In order to envisage internal attributes of an optically opaque object by non-invasive means the fundamental requirement is a form of energy that can pass through the material of which it is made. Imaging methods fall into two categories, namely emission and transmission techniques. In the first of these, if the feature of interest itself emits signals such as gamma-rays from radioactive regions of an object, these can be sensed by external detectors to reveal the position and intensity of the radiation source. More commonly, external beams of penetrating radiation are used and their degree of attenuation as they pass through an object can be used to deduce its density distribution. These general principles are illustrated by considering the case of energetic photons (X- and gamma-radiation).

2.1. Attenuation of High-energy Photons in Matter

A narrow beam of photons of a given energy is attenuated exponentially in uniform homogeneous materials, therefore the intensity of the primary beam after passing through a linear thickness t is given by $I_t = I_0 \exp(-\mu t)$ where I_0 and I_t are the incident and transmitted beam intensities, and μ is the linear attenuation coefficient. For a variable density inhomogeneous medium μ is replaced by $\mu(x, y)$, which gives its value at any point (x, y) in the plane of interest, and the total attenuation factor is got by integrating along the beam; thus $I_t = I_0 \exp[-\int \mu(x, y) ds]$, where ds is a small element of path length along the beam. Taking natural logarithms and inverting gives $\ln\left(\frac{I_0}{I_t}\right) = \int_D^S \mu(x, y) ds$; the line integral is summed along the whole beam path between source S and detector D. The average attenuation coefficient along the beam path is simply this line integral divided by the distance between source and detector. The quantities I_t and I_0 are recorded for each position of the scanning beam to measure these line integrals (ray-sums) at a set of closely spaced steps across each plane of interest. Sets of such ray-sums form a projection, and the image recorded by the detector is simply a map of these projections on the detector plane. These projection images are recorded at a large number of angles around the test object. Radon, in 1917, proved that any three dimensional distribution can be reconstructed from such line integrals, and tomographic reconstruction is based on this principle.

Figure 3a shows how this can be applied to the case of transmission tomography where a collimated external beam passes through the object, and the transmitted beam is detected in a collimated detector; tight collimation eliminates scattered radiation to ensure true exponential attenuation of the primary beam. As shown in Figure 3b an internal gamma-emitting source can be visualized by a collimated detector. Figure 3c illustrates some of the many beam paths employed in a scan so that all regions of the chosen object plane are sampled.

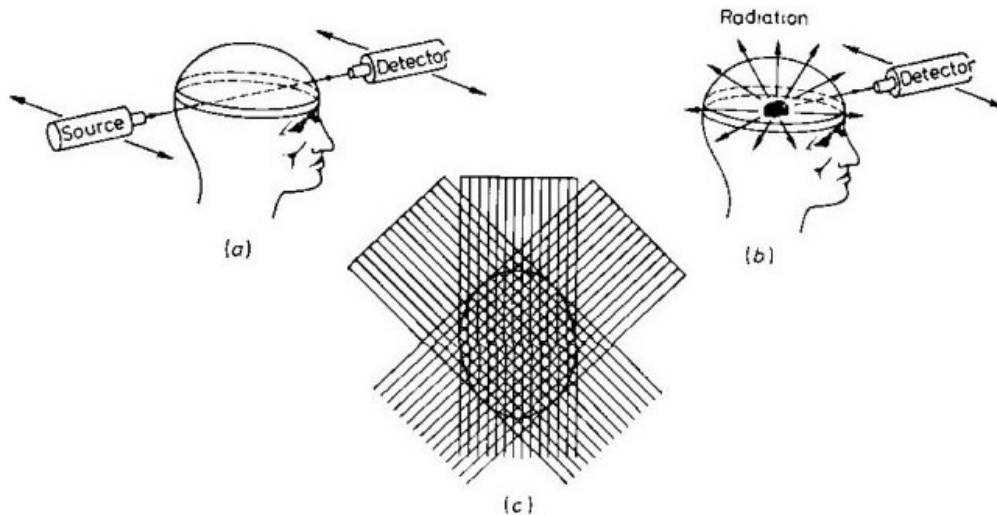


Figure 3: (a) Principles of transmission tomography. A collimated photon source provides an incident beam whose transmitted intensity is measured by a collimated detector. Source and detector are scanned together across the patient to define a narrow trans-axial slice. Sets of such scans are recorded at closely spaced angles around the patient.

(b) Emission tomography: a collimated detector measures gamma radiation emerging from an internal distribution of radioactive material.

(c) Some of the large number of beam paths sampled in a typical CT scan.

In principle, transmission data can be used to construct a large number of simultaneous equations from which the attenuation coefficients of the sub-regions can be determined to yield a two dimensional map of the linear attenuation coefficient distribution in the scanned plane; these values are related to the density of the attenuating medium. In the case of emission tomography a map of the activity distribution is obtained. Since very large numbers of measurements and calculations are usually required, the scanning and reconstruction routines are nearly always controlled by a dedicated computer; it is now customary to describe this technique as computerized tomography, or computed tomography (CT).

2.2. Image Reconstruction

Due to photon statistical variations in CT data, inverting the large matrices to solve simultaneous equations is impractical and a variety of reconstruction routines have been developed which have been briefly reviewed by Gilboy and Foster, 1983, and much more fully by Herman, 1980.

Since this technique measures the average attenuation coefficient along each beam path a first approximation to image reconstruction is to assume that the average value applies equally to all points along the beam. If this so-called simple back-projection method is applied to a large number of projections this results in a $1/r$ blurring of all points in the resulting image where r is the radial distance from each point in the image plane. However this blurring function can be removed to a satisfactory extent by a variety of methods which can be divided into two broad classes, algebraic (iterative) routines and analytical techniques. The first of these is particularly useful where the projection data

are sparse or incomplete, but generally projection images are recorded at closely spaced angles around the test object and analytical methods are in much more common use. Removing a systematic blurring function is termed deconvolution, and a number of techniques based on Fourier transforms have been widely employed. One popular version is called filtered back-projection; this convolves the recorded projections with a function containing negative components before back-projecting them; in effect this subtracts the $1/r$ skirts from all image points when a sufficiently large number of projections are treated this way.

2.3. Image Artifacts

Imperfections in CT images arise from both statistical and systematic sources of error. A fundamental source of error applicable to all modes of radiographic imaging with photons is due to finite photon statistics. Assuming that the scanned cross-section image is reconstructed on an $N \times N$ array of picture elements (pixels) it can be shown that for a given sample, with fixed source intensity and measurement time, the statistical accuracy of the attenuation coefficient assigned to each pixel is proportional to $N^{5/2}$. Thus under these conditions spatial resolution can only be improved at the expense of increased statistical image noise resulting in poorer image contrast. For a given spatial resolution, increasing the source strength, detector efficiency, and measurement time all improve the image contrast to reveal more subtle density variations, but at the expense of increased radiation absorbed dose. In medical applications CT doses in the range 10-100 mGy are typical and CT scanning is one of the highest dose medical radiography procedures. For non-medical CT applications, absorbed dose is not usually a problem, but in the field of micro computer tomography (μ CT) doses can become so large as to raise the question of whether the technique can still be described as non-destructive.

Systematic image artifacts can arise from the limitations of the reconstruction procedure itself and also from practical limitations related to source and detector. Most CT scanning is carried out with an X-ray tube as the photon source mainly on the grounds of its high intensity but the polyenergetic spectrum leads to non-exponential absorption. The softer components of the spectrum are absorbed in the outer layers of the test object, which results in an apparent higher density region in the corresponding parts of the image. In some fan-beam arrangements even small response variations across the face of the detector lead to pronounced ring artifacts that also reduce image quality. Once such sources of instrumental error are understood they can be reduced to acceptable levels.

(see *Limits and Accuracy in Measurements*)

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Biographical Sketch

Walter Gilboy is Senior Lecturer (part-time) in the Physics Department at the University of Surrey in Guildford, U.K. He studied physics at the University of Leeds leading to PhD in 1963. He has worked at Surrey University since 1967 in the field of applied radiation physics and has published over 100 scientific and technical papers. Since 1975 a main interest has been in extending the techniques of medical tomography to non-medical applications mainly with ionizing radiations, but he has also initiated developments in the new field of optical tomography for three-dimensional dosimetry with radiochromic gels. These will be used to verify the complex treatment planning routines being used for conformal radiotherapy. He is a founder member of the International Radiation Physics Society, a Chartered Physicist, and a Fellow of the Institute of Physics.