Preface

Since visible light is reflected by most of the objects around us, our perception of the environment is mainly determined by the properties of their surfaces. To lift this restriction and explore their interior, we have to develop dedicated instruments that rely on penetrating radiation, such as X- and γ-rays, and certain electromagnetic and acoustic waves.

Tomography constitutes the culmination of this endeavor. By combining a set of measurements and performing a reconstruction, it provides a map of a characteristic parameter of the employed radiation for one or more cross sections. It thus enables us to see the interiors of objects on a screen, whereas this was previously only possible either by imagination, based on the measurements, or by direct observation, based on a physical sectioning of the objects. In the case of medical imaging, the latter involved surgical intervention. Tomography is a remarkable invention, which allows us to discover the interiors of the world and the body, their organization in space and time, without destroying them. It is the favored tool for analyzing and characterizing matter, be it dead or alive, static or dynamic, of microscopic or of astronomical scale. By giving access to its structure and the form of its components, it enables us to understand the complexity of the studied object.

Computer aided tomography is a digital image acquisition technique. It produces an encoding, i.e. a digital representation on a computer, of a region of interest inside a patient, a structure, or an object. It thus provides a virtual representation of reality. The digital representation also facilitates subsequent exploitation, exchange, and storage of the associated information. By suitable processing, it then becomes possible to detect the presence of defects, to identify internal structures and to study their form and position, to quantify density variations, to model the components, the body, or the organs, and to guide interventional devices. Moreover, the user may benefit from the assistance of digital image processing, analysis, and visualization software.

A tomographic imaging system comprises several components and technologies. It requires the participation of the final users, such as physicians, physicists, and biologists, for its specification, of researchers and engineers for its development, and of industrial manufacturing and marketing experts for its production and commercialization. These participants are usually educated in medical and engineering schools, as well as at universities.

Throughout this book, we wish to provide help equally to students interested in the scientific and technological background of tomography and to the above mentioned group of people directly involved in the conception and application of tomographic systems. First of all, we focus on explaining the different fundamentals and principles of the formation of tomographic images and on illustrating their aim. Since it is the subject of the series IC2 and of the corresponding English books published by ISTE Ltd. and John Wiley & Sons Inc., we emphasize signal processing and only touch upon the components of the acquisition systems, such as the radiation sources, the detectors, the processing platforms, and the mechanics. Signal processing in tomography forms the intersection between physics for modeling the acquisition systems, mathematics for solving the measurement equations, and computer science for efficiently implementing and executing the image reconstruction. The analysis, visualization, and transmission of these images are addressed in other French books in the series IC2 and the corresponding English books.

This book is compiled from two French books. *La tomographie*[1] corresponds to the first three parts of this book, and *La tomographie médicale*[2] corresponds to the last two parts. For the translation of these two books, several chapters have been updated to reflect advances in the respective domains since their publication.

This book is the result of collective work. Therefore, I would like to dearly thank all authors for their contribution. Tomographic imaging is the "heart" of our work and our research. Each one of us committed him- or her-self to introducing the reader to tomography, in such a way that the origin of the images and the information in these images are comprehensible. By gathering engineers, physicists, mathematicians, and physicians, I formed a multidisciplinary editing team, which allows the reader to benefit from explanations by experts in the respective fields.

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1 GRANGEAT P. (Ed.), *La Tomographie: Fondements Mathématiques, Imagerie Microscopique et Imagerie Industrielle*, Hermes, 2002.

2 GRANGEAT P. (Ed.), *La Tomographie Médicale: Imagerie Morphologique et Imagerie Fonctionnelle*, Hermes, 2002.

The translation of this book was carried out by Holger Eggers. I would like to express my gratitude for his work, which demonstrates not only perfect knowledge of the covered technical aspects but also a good command of both French and English.

In this book, we have compiled in five distinct parts the mathematical foundations associated with image reconstruction, the applications linked to microscopic and industrial imaging, and the applications of medical tomography, separated into morphological and functional imaging.

The book begins with an introduction to tomography, a summary of the domain. This chapter describes the large variety of tomographic systems across the range of accessible contrasts, the choice of acquisition strategies to localize information in space and time, the different approaches to define reconstruction algorithms, and the variety of application domains.

Since the series IC2 and the corresponding English books address signal processing, we have compiled in the first part of the book the mathematical foundations, which serve the development of reconstruction algorithms. Analytical approaches, data sampling, and discrete approaches are discussed successively.

Attempting to cover the applications of tomography exhaustively in a limited number of pages is unrealistic. Therefore, we have selected a set of contributions that illustrate the domain with examples, primarily from French-speaking experts who are actively involved in research in the respective fields. For all chapters devoted to the applications of tomographic systems, we have been committed to describing the physical, physiological, and technological principles that underlie data acquisition and contrast creation. These acquisition strategies lead to the direct problem, which describes the relation between the image to be reconstructed and the performed measurements. The reconstruction algorithms, which attempt to solve the inverse problem, are only mentioned in these chapters, and references are given to the first part of the book for the mathematical derivation. However, the specific problems of each modality, such as pre- and post-processing of data for the correction of parasitic physical effects, are covered in more detail. Finally, several chapters contain sections in which one or several typical applications of the covered imaging modality are described.

The exploration of matter naturally leads to the investigation of smaller and smaller structures, the enhancement of spatial resolution, and the reduction of the scale of the images. Thus, we leaves the dimensions of the human body and looks at samples, cells, proteins, or genes. This is the domain of microtomography. Certain instruments applied in this domain are simply a miniaturization of tomographic systems employed in medical imaging, such as micro CT, MRI, SPECT, and PET

scanners. In this book, however, we are more interested in the instruments that are unique to this domain. The first chapter of the second part of this book is devoted to microscopic tomography and describes in particular the confocal scanning microscope in more detail. The second chapter deals with optical imaging in diffuse media. The last chapter covers tomography with synchrotrons, which are very intense and spatially coherent X-ray sources.

The third part of this book addresses industrial applications of tomography. These must respond to the increasing demands on quality control in manufacturing, on security, and on design. Tomography thus assists design, control and maintenance engineers. In analogy to medical imaging, we have selected a chapter on X-ray tomography for the imaging of containers, which may be associated with morphological medical imaging. This chapter notably describes several examples to illustrate different uses. The second chapter covers emission tomography applied to the visualization of industrial flow, i.e. the imaging of contents, like functional medical imaging.

Medical imaging constitutes the domain in which tomographic systems are developed the furthest. Two parts of this book are devoted to the modalities that are applicable to humans, either for clinical purposes or for cognitive studies. Tomography assists physicians in diagnosis, planning and intervention.

The fourth part of this book deals with morphological medical imaging and covers successively computed tomography, X-ray volume tomography, and magnetic resonance imaging. Since ultrasound imaging is mostly applied to observation of the surface of internal organs, this modality is not addressed in this book. Another book in the series IC2 is devoted to depth imaging, to which ultrasound imaging naturally belongs.

The fifth part of this book covers functional medical imaging in its different forms, namely single photon emission computed tomography, positron emission tomography, functional cerebral tomography by magnetic resonance imaging, and tomography of electrical activity by magneto- and electro-encephalography.

By enabling us to see the invisible, to look inside matter, tomographic systems have a magical, mysterious aspect. They are routinely used tools, which open up the possibility for physicians, researchers, and engineers to answer fundamental questions on the organisms or objects that they examine. Throughout this book, we invite the reader to understand the magic of these tools and thus to discover the exciting world of tomography.

Pierre GRANGEAT

Chapter 1

Introduction to Tomography

1.1. Introduction

Tomographic imaging systems are designed to analyze the structure and composition of objects by examining them with waves or radiation and by calculating virtual cross-sections through them. They cover all imaging techniques that permit the mapping of one or more physical parameters across one or more planes. In this book, we are mainly interested in calculated, or computer-aided, tomography, in which the final image of the spatial distribution of a parameter is calculated from measurements of the radiation that is emitted, transmitted, or reflected by the object. In combination with the electronic measurement system, the processing of the collected information thus plays a crucial role in the production of the final image. *Tomography* complements the range of imaging instruments dedicated to observation, such as radar, sonar, lidar, echograph, and seismograph. Currently, these instruments are mostly used to detect or localize an object, for instance an airplane by its echo on a radar screen, or to measure heights and thicknesses, for instance of the earth's surface or of a geological layer. They mainly rely on depth imaging techniques, which are described in another book in the French version of this series [GAL 02]. By contrast, tomographic systems calculate the value of the respective physical parameter at all vertices of the grid that serves the spatial encoding of the image. An important part of imaging systems such as cameras, camcorders, or microscopes is the sensor that directly delivers the observed image. In tomography, the sensor performs *indirect measurements* of the image by detecting the radiation with which the object is examined. These measurements are described by the radiation transport equations, which lead to what

Chapter written by Pierre GRANGEAT.

mathematicians call the *direct problem*, i.e. the measurement or signal equation. To obtain the final image, appropriate algorithms are applied to solve this equation and to reconstruct the virtual cross-sections. The reconstruction thus solves the *inverse problem* [OFT 99]. Tomography therefore yields the desired image only indirectly by calculation.

1.2. Observing contrasts

A broad range of physical phenomena may be exploited to examine objects. *Electromagnetic waves*, *acoustic waves*, or *photonic radiation* are used to carry the information to the sensor. The choice of the exploited physical phenomenon and of the associated imaging instrument depends on the desired contrast, which must assure a good discrimination between the different structures present in the objects. This differentiation is characterized by its *specificity*, i.e. by its ability to discriminate between inconsequential normal structures and abnormal structures, and by its *sensitivity*, i.e. its capacity for measuring the weakest possible intensity level of relevant abnormal structures. In medical imaging, for example, tumors are characterized by a metabolic hyperactivity, which leads to a marked increase in glucose consumption. A radioactive marker such as fluorodeoxyglucose (FDG) enables detection of this increase in the metabolism, but it is not absolutely specific, since other phenomena, such as inflammation, also entail a local increase in glucose consumption. Therefore, the physician must interpret the physical measurement in the context of the results of other clinical examinations.

It is preferable to use *coherent radiation* whenever possible, which is the case for ultrasound, microwaves, laser radiation, and optical waves. Coherent radiation enables measurement not only of its attenuation but also of its dephasing. The latter enables association of a depth with the measured information, because the propagation time difference results in dephasing. Each material is characterized by its *attenuation coefficient* and its *refractive index*, which describe the speed of propagation of waves in the material. In diffraction tomography, we essentially aim to reconstruct the surfaces of the interfaces between materials with different indices. In materials with complex structures, however, the multiple interferences rapidly render the phase information unusable. Moreover, sources of coherent radiation, such as lasers, are often more expensive. With X-rays, only phase contrast phenomena that are linked to the spatial coherence of photons are currently observable, using microfocus sources or synchrotrons. This concept of spatial coherence reflects the fact that an interference phenomenon may only be observed behind two slits if they are separated by less than the coherence length. In such a situation, a photon interferes solely with itself. Truly coherent X-ray sources, such as the X-FEL (*X-ray free electron laser*), are only emerging.

Part 1

Image Reconstruction

Chapter 2

Analytical Methods

2.1. Introduction

Tomography is a technique for studying matter based on measurements of emitted or transmitted radiation. These measurements are described by equations that link the radiation sources, the interaction of the radiation with matter, and the detectors. The *reconstruction of images* from these indirect measurements involves calculating maps of a characteristic parameter by inverting the measurement equations.

This characteristic parameter is linked to the radiation sources in emission tomography and to the coefficients describing the interaction of the radiation with matter in transmission tomography. Image reconstruction is an *inverse problem*, in which first the *direct problem*, linking the measurements to the characteristic map, is formulated based on physical laws and then the characteristic map is calculated by solving the associated equations.

The *analytical methods* rely on a description of the images and measurements by continuous functions and on a modeling of the physical laws by functional operators. They are of particular interest whenever the inverse operator can be expressed in the form of an explicit inversion formula or a concatenation of such formulas. Their numerical implementation leads to algorithms in which the images are directly calculated from the measurements in a single step, without resorting to more time-consuming, iterative methods.

The production and transport of photons in matter are described by partial differential equations, such as the Boltzmann equation. In *image reconstruction*, the simplified integral form using projection operators is preferably employed. These

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Chapter written by Michel DEFRISE and Pierre GRANGEAT.

integral projections reflect the fact that the contributions of the whole of the traversed matter are accumulated. In general, these equations have the form of Fredholm integral equations of the first kind. The main analytical operators used in image reconstruction are the *Radon transform*, the *X-ray transform*, and the *Fourier transform*, which are all described in this chapter.

The algorithmic implementation of these analytical methods relies on a discretization of the inversion formula. The numerical analysis basically employs filtering, backprojection, and summation operations, as well as discrete Fourier transforms. The need to accelerate the calculation leads to adaptations of the algorithms to the architecture of the computing hardware, use of efficient numerical operators, like the fast Fourier transform, and simplifications of the calculations, for instance by rebinning to other geometries.

Image reconstruction is an *ill-posed inverse problem* in the sense of *Hadamard*. In particular, the reconstruction filters amplify the noise that is present in the measurements. To reduce the influence of these statistical fluctuations, *regularization* techniques are used, which rely on a smoothing by low-pass filters, as discussed in section 2.2.4.

These analytical inversion formulas also enable study of the data sampling, as described in Chapter 3. In addition, they provide explicit formulas for the calculation of expressions that characterize the acquisition systems, as the signal-to-noise ratio in the reconstructed images or the transfer function of the tomographic imaging systems.

The *analytical methods* are very general because they do not rely on characteristic properties of the studied object. However, explicit inversion formulas exist only for simple operators. When complex physical phenomena are introduced into the direct operator, such as multiple scattering in single photon emission computed tomography, an explicit inversion formula no longer exists. Similarly, if *a priori* information, constraints, or selection criteria concerning the images to be reconstructed are to be introduced, the solution is often defined by the minimum of a cost function, for which an explicit formula rarely exists. In these cases, the *discrete methods* described in Chapter 4 are preferably used.

In this chapter, we introduce the principal *operators* that serve the description of different *acquisition geometries* encountered in tomographic systems, most notably the X-ray and the Radon transform, associated with integral projections along lines or over hyperplanes, respectively.

We discuss the two-dimensional (2D) and the three-dimensional (3D) case and distinguish between parallel and divergent geometries, where the latter include the *fan-beam geometry* in 2D and the *cone-beam geometry* in 3D. We summarize the

main properties of these operators and describe the inversion approaches on which the analytical reconstruction algorithms that are used on tomographic imaging systems rely. Then we address 3D positron emission tomography in section 2.6 and X-ray tomography in cone-beam geometry in section 2.7. Finally, we discuss in section 2.8 dynamic tomography where the reconstructed image is changing over time.

For a more detailed description and for application examples of certain formulas introduced in this chapter, the reader is referred to [BAR 81, DEA 83, HER 80, KAK 88, NAT 86, NAT 01, BAR 04], especially for sections 2.2 and 2.3, which deal with the 2D Radon transform. References to the original articles, in which the results presented below were introduced, can be found there as well. The reader may also refer to the following reviews [GRA 01, HIR 97, WAN 00].

2.2. 2D Radon transform in parallel-beam geometry

2.2.1. *Definition and concept of sinogram*

The function $f(M)$ represents the value of the studied physical quantity at a point M. In the 2D case, M is in the plane of the measured slice. The Cartesian coordinates (x,y) of M are used in this plane. Depending on the context, $f(M)$ or $f(x,y)$ is written. The function f is assumed to be sufficiently regular and to be zero outside a centered disk with radius R.

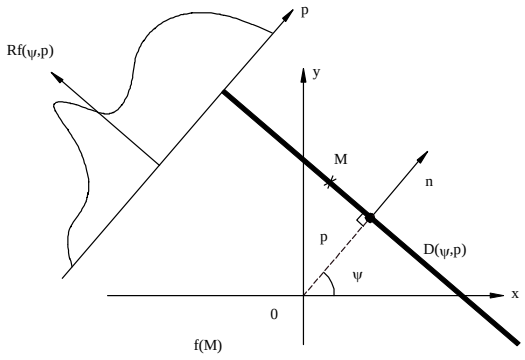

Figure 2.1. *2D Radon transform in parallel-beam geometry (equation [2.1])*

The *2D Radon transform*, or *2D X-ray transform*, associates with the function $f(M)$ the set of its integrals along lines D in the plane of the slice. The lines D are

Chapter 3

Sampling Conditions in Tomography

3.1. Sampling of functions in $\Re^n$

When a signal, or a function, is to be measured, the question arises regarding with which frequency it should be sampled. In this chapter we investigate the geometry of sampling schemes in multiple dimensions. The practical objective of this study is to sample sufficiently densely to obtain a very accurate global estimate, while minimizing the total number of samples taken, to achieve high efficiency. The choice of an adequate regular sampling scheme may be guided by Fourier analysis. We summarize the basic results of multidimensional Fourier analysis, which enable Shannon's sampling conditions to be established, and then apply them to tomography. The reader is referred to [JER 77] for an introduction to Shannon's sampling techniques.

3.1.1. *Periodic functions, integrable functions, Fourier transforms*

We summarize here the basic definitions of periodic functions, square-integrable functions, and their Fourier transforms. For $n \in \mathrm{N}$ and a non-singular matrix $\mathbf{W} \in \Re^n$, the function f from $\Re^n$ to $\Re$ is called periodic with period $\mathbf{W}$ if $\forall \mathbf{x} \in \Re^n, \forall \mathbf{k} \in Z^n$, $f(\mathbf{x} + \mathbf{W}\mathbf{k}) = f(\mathbf{x})$. A set $K \subset \Re^n$ is called a fundamental set associated with $\mathbf{W}$ if K is bounded and if $K + \mathbf{W}\mathbf{k}, \mathbf{k} \in Z^n$ is a tiling of $\Re^n$. In particular, the parallelepiped spanned by the columns of the matrix $\mathbf{W}$ is a fundamental set associated with $\mathbf{W}$ provided that the opposite sides are left open and closed, respectively: ($\left\{\mathbf{y} \in \Re^n, \exists \mathbf{x} \in [0,1[^n, \mathbf{y} = \mathbf{W}\mathbf{x}\right\}$ is an example of a

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Chapter written by Laurent DESBAT and Catherine MENNESSIER.

Chapter 4

Discrete Methods

4.1. Introduction

In this chapter, discrete methods of tomographic reconstruction are described, which are, in contrast to analytical methods, by definition based on discrete modeling of the image to be reconstructed and the measured data. These methods, which are also referred to as series expansion methods, encompass several large classes of techniques. They have been developed in different contexts to better examine certain physical and statistical phenomena that are inadequately described by the Radon transform. These methods permit integration of prior information on the acquisition process and on the images to be reconstructed, and they allow the flexible use of diverse optimization criteria. We distinguish two large classes of techniques. First, we describe the algebraic methods, which are generalized inverse methods that are adapted to a particular form of the Radon operator. The most commonly used ones are those derived from subspace projection techniques. Then, we present the statistical methods, which have been developed in the framework of Bayesian estimation or estimation by functional optimization. They have been elaborated by considering the context of the data acquisition determining the nature of the problem, which can be over- or under-determined. In the first case, the solution is unique, provided that consistency problems do not arise from different types of noise (measurement noise, discrepancy between model and data, etc.). By contrast, in the second case, only a limited number of measurements are available, which does not allow determination of a unique solution. The inverse problem may thus be weakly or strongly ill-conditioned. In the following descriptions of the individual methods they are classified into four categories resulting from the distinction between algebraic and statistical, and over- and under-determined methods.

Chapter written by Habib BENALI and Françoise PEYRIN.

In each section, we outline the main algorithms in use, which are, in view of the size of the problem, generally iterative.

4.2. Discrete models

The discrete methods rely on a discrete representation of both the image to be reconstructed and the measured data.

The image $f(\mathbf{x})$ is represented by the vector $\mathbf{f}$ with coordinates f_j in a finite basis of N square summable functions $h_j(\mathbf{x})$:

$$f(\mathbf{x}) = \sum_{j=0}^{N-1} f_j \; h_j(\mathbf{x}) \qquad [4.1]$$

The most natural decomposition is the choice of the indicator functions of pixels or voxels for the functions h_j. Other choices, such as “natural pixels” [BUO 81], “blobs” [LEW 92], B-splines, or wavelets [GUE 90], are possible as well.

Generally, a projection measurement m_i is expressed as a line integral along a path $s_i(\mathbf{x})$:

$$m_i = \int f(\mathbf{x})\, s_i(\mathbf{x})\, d\mathbf{x} \qquad [4.2]$$

By introducing the decomposition of $f(\mathbf{x})$ and exploiting the linearity of the continuous and discrete summation operators, the measurement is described by:

$$m_i = \sum_{j=0}^{N-1} \left[\int h_j(\mathbf{x})\, s_i(\mathbf{x})\, d\mathbf{x}\right] f_j \qquad [4.3]$$

Regrouping the set of measurements in the vector $\mathbf{m}$ leads to the discrete model in matrix–vector form:

$$\mathbf{m} = \mathbf{R}\,\mathbf{f} \qquad [4.4]$$

where $\mathbf{R}$ it a matrix of size $M \times N$, whose elements r_{ij} represent the i-th projection measurement of the basis function $h_j(\mathbf{x})$.

This formulation is generic because the matrix $\mathbf{R}$, which represents the projection operator, may be defined in parallel or divergent geometry, in 2D or 3D, and for X-ray or emission tomography.

Part 2

Microtomography

Chapter 5

Tomographic Microscopy

5.1. Introduction

Microscopy remains the method of choice for biologists and pathologists for observing cells and biological tissues. The observation is, however, limited to more or less thin sections between 5 µm and 50 µm in conventional photon microscopy, and between 80 nm and 200 nm in electron microscopy. The obtained images represent the information contained in a section and do not allow understanding of the three-dimensional (3D) organization of tissues, cells, and their organelles without ambiguity. Only 10 years ago it was still necessary to physically make series of sections to obtain this 3D information. This extremely painstaking work was followed by the reading of the contours of objects and the 3D reconstruction with adapted software. Besides the delicate and tedious nature of the collection of the sections, the limitations of this technique reside in the impossibility of avoiding distortions of the tissue during microtomy and in the difficulty in defining reliable fiducial markers that permit realignment of a series of sections [BRO 90, DUX 89].

Since the beginning of the 1980s, more powerful methods, which preserve the integrity of the observed specimens, have been developed. They are based on either projection tomography techniques or optical sectioning techniques. The reconstruction methods for projection tomography have essentially been developed in high-energy electron microscopy (section 5.2). While MRI does not currently provide a spatial resolution that meets the requirements of microscopy (see Chapter 12), the recent development of a commercial system for X-ray microtomography now enables resolutions in the order of 10 µm to be achieved. The invention of X-ray microscopy

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Chapter written by Yves USSON and Catherine SOUCHIER.

techniques that use synchrotron radiation will lead to accelerated development of projection methods in high-resolution microscopy, which promise the attainment of resolutions in the order of 25 nm (see Chapter 7). These techniques remain reserved to research laboratories with privileged access to synchrotron radiation.

The techniques of optical sectioning for microtomography [AGA 89] have seen an accelerated development during the last decade. Their success, and in particular that of confocal microscopy (section 5.3), is due to their ease of implementation. These methods directly produce a stack of images that are perfectly aligned and easily interpretable, and the availability of numerous commercial systems facilitates access to them at non-prohibitive costs. In addition, in the domains of biology and medicine, the development of immunofluorescence techniques, of fluorescence *in situ* hybridization (FISH), and of fusion proteins [GIE 06] has substantially contributed to promoting the use of photon confocal microscopy [PAW 06, CON 05].

Confocal microscopy may also be used in the analysis of surfaces. This type of application, which is receiving growing attention in materials science, is addressed in another book in the series IC2 [JOU 02].

5.2. Projection tomography in electron microscopy

Transmission electron microscopy (TEM) is indisputably the technique that enables exploration of tissues and cells with the best possible resolution. Sizes are expressed in tens of nanometers, i.e. in angstroms. To be able to gather consistent 3D information, sections of 1 to 3 µm thickness have to be used. These thicknesses are far higher than those commonly employed in TEM (80 nm to 150 nm). As a matter of fact, the techniques of projection tomography in conventional TEM require microscopes with high energy of the order of 1 MV or more. Below these energies, the penetration capability of the electrons is insufficient for traversing sections of 1 µm and for obtaining images with adequate contrast. Very fortunately, the development of the scanning TEM (STEM) has enabled the use of lower voltages of the order of 300 kV.

The exploration of specimens in 3D with STEM is thus accomplished by acquiring a series of images, between which the observed object is revolved from –60° to 60° in regular angular steps (of 1 to 3°) around the y axis of the plane. In practice, the revolution is carried out by placing the specimen on a goniometric eucentric stage. The volume reconstruction is simply done by applying backprojection algorithms such as ART (Chapter 4). However, the method has two peculiarities compared to the methods for macroscopic projection tomography. The first is the poor control of the scale of the microscope's goniometric stage. In fact, any minor eccentricity entails a notable shift of the projections. So, a systematic realignment of all projections must be performed.

Chapter 6

Optical Tomography

6.1. Introduction

The interaction of light with matter gives access to both the composition and the structure of matter. Optical spectroscopy, i.e. the study of how light is absorbed or emitted by matter, is an important source of information about the composition and the atomic and molecular structure of solid, liquid and gaseous media. In the case of solid, often heterogenous matter, it is frequently necessary to determine the local composition of materials more precisely and to produce for this purpose images of them. In the case of opaque matter, such as metal and materials with high optical absorption, only their surface may be characterized by its interaction with light. Images of the surface may then be made, which, depending on the acquired data, allow more precise determination of its composition, its dielectric properties, its structure, or its texture. This gives insights into "surface states". In the other cases, the light penetrates the material at least partially, and this enables a three-dimensional (3D) image of the composition or structure of the matter to be obtained by studying the dielectric properties. The representation of the latter in a single or multiple cross sections is the result of what is called "optical tomography" methods, which are outlined in this chapter. These methods rely on knowledge of the laws that govern the propagation of light in matter. These laws vary, depending on whether the matter is transparent or only translucent, i.e. whether the light penetrates the matter but loses the property of ordered or coherent propagation due to frequent encounters with heterogenities along its trajectory. The laws that govern the propagation may in general be modeled mathematically or simply numerically using computers. This is the "direct" model, which is of general relevance to tomographic methods. However, there is also a problem specific to the domain of electromagnetic wave optics, in which the coherent

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Chapter written by Christian DEPEURSINGE.

and incoherent propagation of waves may simultaneously be observed in varying ratios. Solving the "inverse" problem is far more complicated, and the precision reached depends on whether a perfectly transparent or only a translucent medium is analyzed, or a turbid medium, in which the scattering of light by the heterogenities of the matter becomes a dominant factor.

An application domain of growing importance for optical tomography is the observation of biological matter. The problem of "mixed" propagation of coherent and incoherent waves acutely arises in this case because of the particular characteristics of the structure of biological media. These are composed of comparatively well- structured, repeating units, of which namely the organelles in cells are identifiable by their size and dielectric properties. This property is very useful in the differentiation of tissues and serves as the basis for optical diagnostics exploiting absorption and scattering properties of light.

To supplement the reading of this chapter, a certain number of references are recommended, which deal with:

– the propagation of electromagnetic waves in a vacuum and dielectric materials [JAC 95];

– the principles and theories concerning coherent optics [BOR 99];

– the tomographic techniques applied in optics [BOR 99, KAK 88];

– the interaction of light with living matter, the optical absorption and scattering, and thermal balance [WEL 95];

– the physical properties of tissues [DUC 90].

6.2. Interaction of light with matter

Generally speaking, light interacts with matter, mainly with the charged particles, such as electrons and protons, according to the well-known coupling between electromagnetic fields and electric charges (Maxwell equations, see [JAC 95]). Indirectly, it also interacts, via the charged particles, with the atoms, molecules, and crystal lattices, or more precisely with their vibrational modes. We will describe this interaction in more detail, notably in the framework of organic and living matter. In principle, two types of interaction are distinguished: absorption, often linked to the phenomenon of fluorescence, and optical elastic or inelastic scattering. Basic references on the interaction of light with matter are countless. We therefore recommend readers to refer to elementary references [BOR 99]. The following basic references are also very useful in understanding this section

Chapter 7

Synchrotron Tomography

7.1. Introduction

Synchrotron radiation is a very bright source of X-rays with a very broad energy spectrum. It is produced by a relativistic electron beam deflected by strong magnetic fields. The photon beam may be made monochromatic, usually with crystals, while preserving a considerable photon intensity. Synchrotron radiation thus provides an intense, monochromatic X-ray beam with very small divergence and source size. It is well suited to serve as a source in quantitative tomography and microtomography, as well as in more exotic applications, such as fluorescence, diffusion and phase contrast tomography. In section 7.2 we describe the principle and advantages of synchrotron radiation. We discuss quantitative tomography in section 7.3 and microtomography in section 7.4. Finally, we briefly describe in section 7.5 the experimental efforts put into other applications: phase contrast, holographic, refraction, diffraction, diffusion, and fluorescence tomography. We restrict ourselves to biomedical applications. However, synchrotron radiation is particularly well suited for tomography on non-biological objects, where the constraints on dose and acquisition time are relaxed (see Chapter 8).

7.2. Synchrotron radiation

7.2.1. *Physical principles*

A relativistic electron beam deflected by a magnetic field emits bremsstrahlung, i.e. a deceleration radiation, called synchrotron radiation. This phenomenon was observed

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Chapter written by Anne-Marie CHARVET and Françoise PEYRIN.

for the first time with a small synchrotron at the research laboratory of General Electric in 1947 (Schenectady, NY, USA). This radiation is very intense, directional, and generally covers a broad range of energies. Therefore, it is of interest as a source of ultraviolet (UV) and X-rays. Its characteristics depend on both the magnetic field and the electron beam.

When the magnetic field is uniform and static, electrons describe a circular trajectory in a plane perpendicular to the magnetic field. The photons are emitted in a cone tangential to the trajectory of the electrons. For the user, the UV and X-rays appear as a continuous layer in the horizontal plane with a very small vertical opening angle. The electromagnetic radiation emitted in such a magnetic field – called dipolar – has a very broad, continuous energy spectrum, which extends from radiofrequencies to UV and X-rays. This spectrum is characterized by an energy, called the critical energy, which is defined as the value below which half of the power is emitted. This critical energy ε_c (measured in keV) depends on the energy of the electrons E (in GeV) and on the strength of the magnetic field B_0 (in T):

$$\varepsilon_c = 0.665 E^2 B_0 \qquad [7.1]$$

At the ESRF (*European Synchrotron Radiation Facility*, Grenoble, France), the critical energy is 19.2 keV for the bending magnets. The vertical opening angle of the X-ray beam depends on the energy of the photons. The photons with the highest energy are emitted in the cones with the smallest opening. At the ESRF, this is 170 μrad at the critical energy, which corresponds to an X-ray beam of 1.7 mm in height at 10 m from the source. For a dipole magnet, the horizontal opening is defined with the help of slits. Due to the very anisotropic emission of synchrotron radiation in the vertical direction, the appropriate quality factor is the, often vertically integrated, spectral intensity. It is measured in photons per second per milliradian (horizontal) and per 0.1% $\Delta\lambda/\lambda$. The last quantity means that the energy band used for the calculation for each λ is $\Delta\lambda = 0.1\%\ \lambda$.

The use of periodic magnetic structures allows changing of the characteristics of synchrotron radiation, in particular its energy spectrum and its angular emission, by an interference phenomenon of the radiation emitted by successive magnetic periods. In the simplest case, these structures consist of magnets of alternating polarity that result in a sinusoidal trajectory of the electrons. They are often referred to by the term insertion devices and are characterized by a parameter K, called the deflection parameter:

$$K = 0.934\ B\ \lambda_u \qquad [7.2]$$

Part 3

Industrial Tomography

Chapter 8

X-ray Tomography in Industrial Non-destructive Testing

8.1. Introduction

The first images obtained with tomography in the framework of process monitoring (process tomography) [BAR 57] enabled mapping of the density of particles in suspension in a fluidized bed. Industrial applications diversified after Hounsfield developed the first medical scanner in 1972 [THI 98]. Today, tomography allows the detection of defects, dimensional control, characterization of materials (e.g. measurements of density and distribution of impurities), and approval of new manufacturing processes. With the evolution in computer science and CCD cameras, three-dimensional (3D) tomography has become of primary importance for modeling the behavior of materials, for instance the damage within advanced composites or the fluid flow within rocks. The size of objects may vary from a millimeter to several meters and the size of voxels from less than a micrometer to several centimeters. It should be mentioned that all types of materials may be tested (including metals, composites, and ceramics). The term tomography is often used to designate transmission tomography, which is employed most and is based on the measurement of the attenuation of photons through an object. Yet the same term covers other types of imaging, notably emission, fluorescence, and scattering tomography. The last, which is based on the analysis of photons that are scattered by an object (by the Rayleigh and/or Compton effect), is mentioned at the end of this chapter (see section 8.6.9).

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Chapter 9

Industrial Applications of Emission Tomography for Flow Visualization

9.1. Industrial applications of emission tomography

9.1.1. *Context and objectives*

The understanding of flow phenomena in general, and multiphase flow phenomena in particular, is crucial in the development and control of processes in industry and process engineering. Flow systems with fluids in two or more non-miscible phases are found in numerous branches of industry, including the oil, food, pharmaceutical, and materials (foundry, plastics) industries.

To illustrate the subject of this chapter, we consider the problem of designing a reservoir that is agitated by a chemical reactor. The objective of the designer is not only to optimize the system in economical terms, including its dimension and its production, but also to develop a complete control system, which assures efficient operation of the process. For this purpose, the designer resorts to mathematical models constructed from the mass and energy balance, the kinetics of the chemical reactions, and properties of the materials. However, it is often impossible to make predictions from basic principles alone. Despite a variety of processes that use multiphase flows, the understanding of the hydrodynamic problems encountered remains incomplete. The flow patterns distinguished are annular, laminar, bubble and turbulent. The designer of the reactor thus has to rely on experimental data, acquired in earlier experiments in the laboratory, or in tests on a pilot installation at reduced scale.

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Part 4
Morphological Medical Tomography

Chapter 10

Computed Tomography

10.1. Introduction

10.1.1. *Definition*

Computed tomography (*CT*) is a morphological imaging modality that maps the density of human tissue in cross-sections. It employs X-rays, generated by an X-ray source, to collect the required information. The X-ray source and a linear detector, composed of a set of detector elements, are mounted opposite each other on a gantry. Both are rotated around the patient, which determines the examined cross-section [CAR 99, NEW 81, ROB 85].

The phenomenon that enables recovery of the distribution of the density is the attenuation of the X-rays as they pass through the absorbing human body. The examined cross-section is described by a function $\mu(x,y)$, which represents the linear attenuation coefficient of the tissue (for a given energy, μ is proportional to the density).

On the assumptions that radiation with energy E is monochromatic and that the beam with incident flux ϕ_0 is parallel and infinitely narrow, the transmitted flux of X-ray photons is given by Lambert–Beer's law:

$$\phi = \phi_0 \, \mathrm{e}^{-\int_A^B \mu_E(x,y)\,\mathrm{d}l}$$

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Thus, $-\log \frac{\phi_0}{\phi} = \int_A^B \mu_E(x,y)\,\mathrm{d}l$ corresponds to the integral of the function μ_E along the half-axis from the source through a detector element. Measuring these integrals enables reconstruction of the image $\mu_E(x,y)$. For the mathematical background on image reconstruction from line integrals, the reader is referred to Chapter 2.

After reconstruction, the image is normalized to *Hounsfield units*:

$$H = 1000\ \frac{\mu - \mu_{water}}{\mu_{water}}$$

where μ and μ_{water} are the linear attenuation coefficients of the considered tissue and of water, respectively.

10.1.2. *Evolution of CT*

The impact of CT on *diagnostic imaging* practice was so profound that its inventors, Hounsfield and Cormack, received the Nobel prize in 1979. Since its introduction, CT has advanced considerably. Several generations of scanners have been developed to acquire the data required for image reconstruction more and more efficiently, i.e. to scan the required set of source–detector positions faster and faster.

In the first and second generation *scanners*, the limited number of detector elements made several types of movements of the source and detector necessary for this purpose.

In modern scanners, one of the two following geometries are employed (see Figure 10.1):

–R/R geometry: a divergent beam of radiation is emitted by the X-ray source and irradiates a curved multidetector (composed of in the order of 1,000 detector elements), whose size allows at least the whole examination volume to be covered. The source and the detector rotate together;

–R/S geometry: a multidetector completely surrounds the patient (ring-shaped detector), and only the source rotates around the patient. In particular, so-called electron beam tomography scanners, which completely integrate the X-ray source, rely on this geometry. The position of the radiation source is in this case determined by the direction of the electron beam that generates the X-rays when decelerating in the target. Using this geometry, mechanical movements are no longer performed [MCC 95].

Chapter 11

Interventional X-ray Volume Tomography

11.1. Introduction

11.1.1. *Definition*

Three-dimensional (3D) *X-ray tomography* is a medical imaging modality providing information on the distribution of the density of human tissues within a given volume. The technique is based on the measurement of the attenuation of an X-ray beam passing the subject. The principle of the acquisition is very similar to that in computed tomography (see Chapter 10). Differences result mainly from the system architecture. Instead of using a dedicated tomographic closed bore imaging system, 3D X-ray tomography data are acquired on a conventional X-ray system, in which the X-ray tube and the two-dimensional (2D) planar detector are mounted opposite to each other on a C-shaped support (gantry). For tomographic imaging, the source–detector combination rotates around the patient and determines the volume to be imaged. Reconstruction of the 3D attenuation maps $\mu_a(x, y, z)$ is performed similarly to X-ray computed tomography using the set of measured line integrals of μ_a acquired in cone-beam projection geometry. The mathematical background of the data reconstruction of volumes from measured integrals is described in Chapter 2.

The tomographic imaging approach described in this chapter is mainly focused on *interventional imaging*. The increasing complexity of minimal invasive procedures requires the availability of high resolution 3D image information for intervention

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planning, guidance and outcome control. In this context, recent image data are a key component to allow for accurate guidance during intervention.

11.1.2. *Acquisition systems*

The 2D X-ray detector integrated in an interventional system is usually a *flat detector* device. It is based on an amorphous silicon plate covered with a scintillation material [SCH 94, BUS 02]. Recent detectors cover a planar region of up to 30 x 40 cm at high spatial resolution in the order of 180² µm² pixel size. The maximum frame rate is 60 frames per second.

Similar to classical computed tomography, in an early system for generating volume information for interventional purposes, a ring-shaped gantry was used to obtain high geometrical stability. At that time, image intensifier tubes were used to acquire digital images at high frame rates. This combination has been realized in the morphometer [SAI 93, HEA 95], which has remained a prototype (see Figure 11.1).

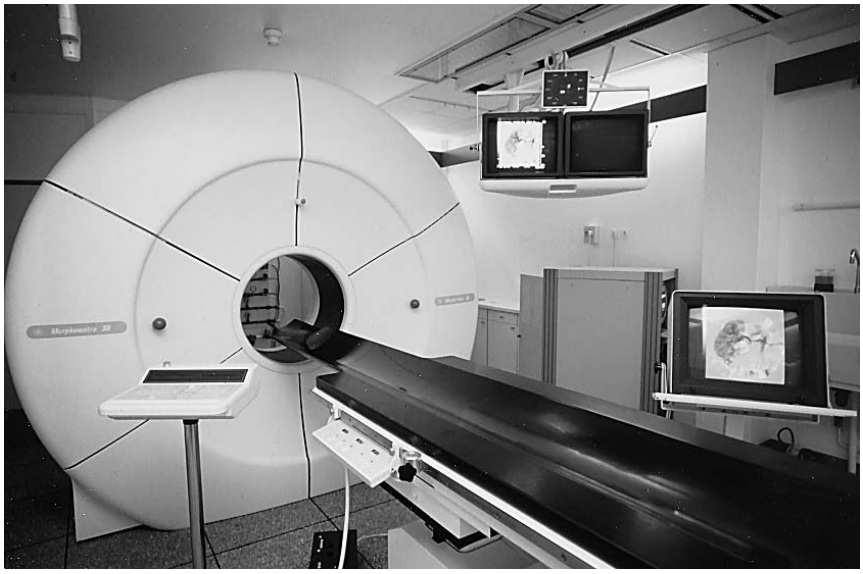

Figure 11.1. *3D scanner: the morphometer (reprinted with kind permission from Springer Science + Business Media, Figure 1 in [HEA 95])*

Chapter 12

Magnetic Resonance Imaging

12.1. Introduction

The importance that magnetic resonance imaging (MRI) has gained over the last 25 years, the extent and diversity of its applications, the peculiarity of the signal formation, as well as the variety of imaging sequences, make it difficult to summarize this modality in a few pages. Inevitably, countless questions have to remain unanswered. The interested reader will find more extensive information on the subject in the book by Haacke *et al.* [HAA 99]. The objective of this summary is to introduce three aspects that are very specific to MRI. The first aspect concerns the physical nature of the underlying phenomenon, which has its origin in the magnetism of nuclei. The resulting precession of the magnetization permits a simple coding of the spatial position by means of the generalized phase of the nuclear magnetic resonance (NMR) signal. The second aspect concerns the collection of the information needed for the image formation. MRI is a digital technique and thus allows direct acquisition of samples in the so-called k-space, the reciprocal space of the image. The reconstruction by discrete Fourier transform immediately enables assigning to each pixel of a gray value that is proportional to the local density of the nuclear magnetization. The third, astonishing, aspect stems from the variety of possible contrasts. The measured magnetization may be modified by numerous physical factors. This is exploited by an appropriate choice of the imaging sequence. The presentation of these principles will be followed by a brief discussion of the volumetric nature of the provided information.

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Part 5
Functional Medical Tomography

Chapter 13

Single Photon Emission Computed Tomography

13.1. Introduction

13.1.1. *Definition*

Single photon emission computed tomography (SPECT) is a functional medical imaging modality that enables non-invasive estimation of the three-dimensional (3D) distribution of a radiopharmaceutical *in vivo*. The tomographic images are reconstructed from multiple projections, or planar images, which are acquired with a gamma camera that rotates around the patient. They enable qualitative analysis (detection of lesions) as well as a quantitative analysis (measurement of the concentration of radiopharmaceuticals in organs or estimation of functional parameters). The basic principles of SPECT were established by Kuhl in 1963 without using a gamma camera or a computer [KUH 63]. Anger, the inventor of the gamma camera, demonstrated in 1967 the feasibility of SPECT with a gamma camera [ANG 67]. The reconstruction of SPECT images with a computer was reported in 1971 [MUE 71], and commercial SPECT systems appeared in 1978. Today, many SPECT systems are combined with computed tomography (CT) systems and thus permit the joint acquisition of anatomical and functional information [SEO 08].

13.1.2. *Functional versus anatomical imaging*

SPECT images reflect the spatial distribution of a radiopharmaceutical whose uptake in the body certainly depends on the anatomy of the involved organs

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(locations, volumes), but is mainly determined by their function, which motivates the use of the term “functional imaging”. The influence of the physiopathological state of examined organs on the obtained images is far weaker in other medical imaging modalities, such as CT, MRI, and ultrasound, which achieve a better spatial resolution and are classified as “anatomical imaging” modalities.

The indications for a SPECT scan are different from those for anatomical imaging modalities and concern all pathologies (cardiac, neurological, oncological, osteoarticular, pulmonary, etc.). In this chapter, we study as an example a myocardial perfusion scan, which is currently a common application of SPECT.

SPECT imaging relies on three components: the radiopharmaceutical whose biodistribution is studied, the gamma camera, which enables detection of the signal emitted by the radiopharmaceutical, and the tomographic reconstruction, which allows estimation and visualization of the 3D distribution of the radiopharmaceutical in the body. These three components are described in more detail in the following.

13.2. Radiopharmaceuticals

The radiopharmaceutical or radioactive tracer is the basis for SPECT imaging, since the studied function depends on it and it is the origin of the measured signal. It combines a vector, which ensures functional selectivity, with a radioactive marker, which emits the photons that enable the localization. In certain cases, the vector and the marker are the same, since the radioactive element itself has an interesting specific biological behavior (^{201}Tl, ^{123}I, ^{131}I).

13.2.1. *Vectors*

Different types of vectors allow study of different physiopathological functions. For example, lung perfusion is studied with radioactively labeled albumin aggregates with a diameter of 10 to 40 μm. After intravenous injection, these are mechanically blocked in the lung capillaries, whose cross-section is smaller. Many tracers enable evaluation of the perfusion of other organs. They leave the vascular space, and they are taken up by the cells of the studied organ in amounts proportional to its perfusion (see section 13.5 on myocardial SPECT). The selectivity of other vectors results from their integration into a metabolic process. For example, polyphosphates are incorporated into the physiological process of bone renewal and thus allow imaging of the skeleton. Molecules that are hormonal precursors are also available, such as methyl iodide–benzylguanidine, which specifically enables study of tumors that produce catecholamine in excess.

Chapter 14

Positron Emission Tomography

14.1. Introduction

14.1.1. *Definition*

Positron emission tomography (PET) is a medical imaging technique that enables us to obtain a three-dimensional (3D) map of a physiological parameter, such as glucose metabolism, blood flow, the receptor density of a neural transmission system, *in vivo* in human organs. This 3D map is derived from a dynamic measurement of the volumetric distribution of a specific *radiopharmaceutical* that is injected into the subject. The first PET scanner was built at the beginning of the 1960s by Rankovitch *et al.* It was constructed of a ring of sodium iodine detectors [RAN 62]. The first computer-assisted scanner was described in 1975 [TER 75, PHE 75].

14.1.2. *PET versus other functional imaging techniques*

The other functional imaging technique that uses radiopharmaceuticals is *single photon emission computed tomography* (SPECT). It differs from PET in how the molecule that is injected into the patient is marked and in the characteristics of the associated detection system.

PET uses positron emitters as markers. These markers have a short half-life of between 2 and 109 min; this requires their production by a cyclotron and their

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incorporation into the radiopharmaceutical at the scanner site for isotopes with the shortest half-lives. SPECT uses gamma emitters as a marker. The shortest half-life of the isotopes used is 6 hours. Moreover, these isotopes are easily available using a generator. Thus, unlike PET, SPECT does not require a *cyclotron* in the vicinity of the imaging center.

The sensitivity of PET is about 100 times higher than the sensitivity of SPECT, because the physical *collimators* are replaced by electronic collimators. In addition, elimination of the physical collimators enables a uniform spatial resolution over the entire field of view to be obtained, instead of a spatial resolution that varies with the distance to the collimator.

Mainly due to the cost of tracer production, PET remained for a long time a dedicated tool for clinical research. Its success was based on the large variety of available radiopharmaceuticals, which results from the simple integration of markers into biological molecules in PET. This enabled the use of PET for validating new therapeutics, such as the transplantation of neurons that are able to supplement dopamine production by the *dopaminergic system* in Parkinson's disease [COC 03]. In section 14.4.2 we briefly describe how PET allows *in vivo* imaging of the dopaminergic transmission system. PET has also become the tool of choice for the imaging of gene expression [TAV 98].

PET enables us to perform a functional mapping of the brain by comparing the measured blood flow in cerebral regions during a motor, sensor, or cognitive stimulation to that at rest [FOX 84]. Such activation studies are at present mostly performed by two other functional imaging techniques, functional magnetic resonance imaging (fMRI) and magneto-encephalography (MEG), which have a better temporal resolution than PET.

Since the mid 1990s, PET imaging of glucose metabolism has been highly successful as a tool for the detection and localization of tumors and for the follow-up of patients after oncological treatment. The interest in PET in oncology is linked to the fact that cancer cells have a higher glucose metabolism than normal cells [PAU 98]. In this way PET provides complementary information to that obtained by radiological studies (CT and MRI). Today, there are in France alone more than 30 imaging centers equipped with PET scanners dedicated to the clinical investigation of glucose metabolism in oncology and more than 10 imaging centers equipped with a cyclotron. Most of them are used part-time by industrial distributors of radio-pharmaceuticals that have clearance for market introduction.

Chapter 15

Functional Magnetic Resonance Imaging

15.1. Introduction

In the past two decades, there have been impressive advances in *functional cerebral imaging* [ORR 95]. Positron emission cameras enable the entire brain to be explored with sufficient sensitivity and spatial resolution to allow a detailed mapping of various biochemical processes or cerebral perfusion. In addition, the dynamics of cerebral activation may be followed with a temporal resolution of the order of a millisecond with electrical or magnetic source imaging.

The advent of functional MRI (*fMRI*) in 1991 [BEL 91] and the ensuing developments in this field [MOO 99] have created an unprecedented interest in functional cerebral imaging within the neurological and cognitive science communities. The technique provides access to hemodynamic information of the same nature as that obtained with *positron emission tomography* (*PET*) using ^{18}O-labeled water as the radioactive tracer of perfusion. Furthermore, it has several characteristics that form the basis for the interest in this technique. It allows exploration of the brain with excellent spatial and temporal resolutions [MOO 99]. It provides functional information that may easily be superimposed on anatomical images with high spatial resolution, which are obtained in the same *MRI* examination. The technique is strictly non-invasive, thus allowing a repetition of examinations at will on healthy subjects. Finally, it requires the use of a whole-body MRI scanner, a tool that is widely available at an operational cost far below that of a positron emission camera, which was previously considered to be the gold standard in functional cerebral imaging.

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These methods for functional imaging enable study of the human brain at a systems level. In this sense, these methods are fundamentally different to electrophysiological methods employed to examine brain function in non-human primates. In particular, they allow identification of the functional specialization of certain cerebral regions for particular cognitive functions which cannot be studied in non-human primates. An example is the involvement of the prefrontal cortex in different speech processes. In addition to contributing to our knowledge about healthy brain function, these methods are obviously also of clinical interest. This is the case for the presurgical determination of the hemispheric predominance of language in epileptic patients resistant to medical treatment or for the presurgical assessment of patients with an intra-cerebral tumor.

15.2. Functional MRI of cerebrovascular responses

The existence of a coupling between cerebral activation and the *cerebrovascular* system has been known for more than a century [ROY 90]. However, the precise mechanisms at the basis of this coupling have still not been fully identified. Nevertheless, it has been established that the activation of a population of neurons generates a local increase in *cerebral blood volume*, *cerebral blood flow*, and *blood oxygenation*. It has also been established that the coupling between neural activation and these cerebrovascular parameters is relatively rapid (within some seconds). In functional cerebral imaging with *PET*, it is the coupling between neuronal activation and cerebral blood flow that is exploited. In *fMRI*, the coupling between neuronal activation and each of the three cerebrovascular parameters may contribute to the measured signals [OGA 98].

The first functional MRI experiments showed local increases in *cerebral blood volume* [BEL 91]. A bolus with exogenous *paramagnetic tracers* (chelated molecules of gadolinium) was injected intravenously, which enables its first cerebral transit to be followed. The cerebral blood volume may be determined (except for a proportionality constant) from signal intensity curves during this transit. Cerebral blood volume maps measured under different experimental conditions may thus be compared using this approach. However, the temporal resolution, with which the experimental conditions may be changed, is low (in the order of several minutes). It is similar to that obtained with techniques using radioactive tracers, by which this approach is directly inspired. In addition, the number of experimental conditions is limited by the small number of allowed contrast agent injections.

Two developments in MRI then started a true revolution in the domain of functional cerebral imaging.

Chapter 16

Tomography of Electrical Cerebral Activity in Magneto- and Electro-encephalography

16.1. Introduction

Functional cerebral imaging comprises the techniques that allow study of the working brain. Its goal is to answer two fundamental questions: where and when do the different information processing steps take place in the brain during cognitive tasks and sensorial stimulation? Among the functional cerebral imaging techniques, the metabolic imaging methods, like SPECT (see Chapter 13), PET (see Chapter 14), and fMRI (see Chapter 15), and the electrical imaging methods, like magneto-encephalography (MEG) and electro-encephalography (EEG), are noteworthy. The former have a high spatial resolution in the order of millimeters, but can observe cerebral phenomena only with a temporal resolution in the order of seconds, or even minutes in the case of SPECT and PET. By contrast, the latter have an excellent temporal resolution in the order of milliseconds, since they directly measure the electrical activity of the brain. Moreover, they are totally non-invasive.

In recent years, the instrumentation in MEG and EEG has considerably advanced, and systems with large numbers of detectors, which enable measurement of the magnetic field around the whole head, are today offered commercially. However, the methods for reconstructing the electrical cerebral activity from the fields and potentials measured on the surface of the head face numerous difficulties. A fundamental limitation is that the reconstruction problem is underdetermined and therefore multiple solutions exist. The first methods that were developed and employed when only small numbers of detectors were available localized a limited number of active regions.

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