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Introduction

The first computed tomography (CT) system was built by Godfrey Hounsfield in 1971. A few years earlier, his co-inventor Allan McLeod Cormack had used the Radon transform and its inverse to theoretically describe a radiological x-ray scanning machine and image reconstruction method. For their research, Cormack and Hounsfield received the 1979 Nobel Prize in Physiology or Medicine.

Currently, CT is a widely used x-ray scanning technique. In its prominent use as a medical imaging device, CT serves as a workhorse in many clinical settings throughout the world. It provides answers to urgent diagnostic tasks such as oncology tumor staging, acute stroke analysis, or radiation therapy planning. Moreover, CT systems are also used in the quality analysis of industrial products or for security screening of luggage at airports.

Spectral CT was introduced in 1975 as an improvement in measurement technology. CT devices were enabled to gain information on the energy-dependent (i.e., spectral) attenuation properties of the object. Different technical realizations were employed. The straightforward solution is the so-called dual-kVp technique. Two CT scans with different x-ray tube acceleration voltages are performed. The two resulting data sets contain information on the spectral x-ray attenuation characteristics of the object. These spectral data can be used to obtain additional information on the object.

The corresponding spectral CT algorithms have two fundamentally different targets. First, two patient tissue types, such as bone and iodine-contrast-media-filled blood, can produce the same range of attenuation gray values in a CT image. In order to differentiate between the two materials, spectral CT data can be weighted for an optimum contrast-to-noise ratio. This task is readily solved for two specific tissues types. The general solution is achieved when the spectral measurement channels are summed with the inverse cube of their mean detected energy as scaling weights.

Alternatively, spectral CT algorithms can provide absolute and quantitative information on the scanned object, e.g., its chemical composition. It is important to understand how spectral CT differs from standard single-energy CT scanning. With the latter, the resulting images consist of water-normalized effective x-ray attenuation coefficients given in Hounsfield unit (HU) numbers. A value of −1000 HU represents air, and a value of 0 HU indicates water. In medical imaging, the absolute HU numbers are only used in a limited number of diagnostic fields, e.g., calcium scoring in coronary heart disease assessments. This is due to the dependency of the HU numbers on the measurement system properties such as the x-ray tube spectrum, the CT reconstruction process, and the detector response function. In comparison, quantitative spectral CT algorithms aim to provide absolute material parameters such as concentrations, densities, or atomic numbers. These parameters can be compared to ground truth values in terms of their accuracy and precision. As a natural limit, x-ray measurement noise and systematical errors challenge the quantitative characteristics of the output data.

In 1976, Alvarez and Macovski (see Ref. 1 in Chapter 3) proposed the first and most successful approach to date, the basis-material decomposition (BMD) algorithm. This algorithm transfers the well-known principal component analysis to x-ray physics. As a result, images of basis-material coefficients (e.g., the distribution of bone and soft tissue in a patient) are obtained. The BMD algorithm has been used as a standard method in scientific research for the last few decades.

However, several practical problems were observed. The technical limitations of early dual-kVp CT scanners in terms of temporal resolution and measurement precision resulted in movement artifacts and limited accuracy of the algorithmic input data. Moreover, the algorithms themselves were affected by noise amplifications and model mismatch that led to a limited overall precision and accuracy. For a combination of these reasons, the method did not enter clinical routine CT imaging at that stage.

In 2006, dual-source CT (DSCT) was introduced as a CT measurement technique in which a second x-ray tube and detector pair was integrated into the CT system. DSCT allows for simultaneous data acquisition at two different x-ray tube voltages. The spectral separation between the two measurement channels can be improved by additional tube prefiltration. In the years following the introduction of DSCT, kVp switching of the x-ray tube and detector-based prototype spectral CT systems were developed as alternative system approaches. These rapid advances in spectral measurement technologies have driven a renewed interest in spectral CT algorithms.

To achieve a structured approach to the fundamentals and clinical applications of spectral CT, this book is divided as follows: In the first chapter, we consider the main clinical motivations for spectral CT applications. In Chapter 2, the measurement properties of spectral CT systems are described. Chapter 3 provides an overview of the current state of research on spectral CT algorithms. Based on this overview, we evaluate the technical realization of spectral CT systems in Chapter 4. Device approaches such as DSCT, kVp switching, and energy-resolving detectors are compared. Finally, Chapter 5 summarizes various algorithms for spectral CT reconstructions and spectral CT image postprocessing, and links these algorithms to clinical use cases.

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