Guest Editorial
Multirow Detector and Cone-Beam Spiral/Helical CT

SPIRAL/HELICAL multirow detector computed tomography (MDCT), also referred to as multislice spiral CT (MSCT), has attracted a major interest since its recent introduction [18], [33]. Compared with single-row-detector systems, MDCT scanners allow faster data collection and thinner slices that support more demanding clinical applications and present new research opportunities. This special issue consists of 12 original papers from this rapidly expanding field. Although MDCT generally refers to commercial CT scanners used by radiologists, the work in this issue is also applicable to emission CT as well as nonmedical applications of X-ray CT, including nondestructive testing and microtomography. In this editorial, we provide a summary of the special issue, an overview of the field, and suggestions for future work.

I. Content of the Special Issue

The first eight papers represent recent work on X-ray CT image reconstruction in multirow detector or cone-beam geometry. The first two papers by Schaller *et al.*report an adaptive axial interpolation theory for a MDCT scanner with experimental verification [13], [29]. Adaptive axial interpolation provides increased flexibility for image reconstruction in clinical practice. By keeping slice thickness and image noise independent of pitch, the MDCT parameter selection is simplified. The paper by Proksa *et al.* introduces the n-PI-method for spiral/helical cone-beam CT (CBCT) [27]. This method assumes a detector array shaped by a helix, permits variable pitch, and is amendable to both exact and approximate reconstruction. The paper by Hsieh addresses the degradation in image quality of spiral/helical MDCT when the gantry is tilted to produce oblique sections [17]. He formulated a model as an analytical basis and developed several compensation schemes. The paper by Bruder *et al.* describes two different approximate single-slice algorithms for CBCT: multirow Fourier reconstruction (MFR) and advanced single-slice rebinning (ASSR), in the framework of the generalized parallel projection using z-filtering [3]. Their studies contain data valuable for the design of spiral/helical CT scanners with medium cone-beam angles. The paper by Kachelrieß *et al.* proposes MDCT algorithms for cardiac imaging [19]. Based on ECG signals, the two dedicated cardiac reconstruction algorithms improve image quality compared with the standard reconstruction algorithms. The paper by Kudo *et al.* [21] deals with the long-object problem; that is, to reconstruct a region of interest (ROI) of a long object from data collected along a helical path that extends only marginally above and below the ROI. Their quasi-exact reconstruction algorithms require no additional circular scans and are in the filtered backprojection format. The paper by Zhao and Wang contains a wavelet-based derivation and construction of the Feldkamp-type algorithms [46]. They found that a three-dimensional (3-D) ROI can be reconstructed without severe artifacts nor any significant bias.

The subsequent papers are on detector evaluation and system characterization. The paper by Hsieh *et al.*presents the signal decay and afterglow characteristics of a solid-state detector, the resultant image artifacts, and a correction scheme [17]. The paper by Fuchs *et al.*compares Xenon and solid-state detectors under working conditions [12]. The paper by Ning *et al.* characterizes a flat-panel-detector-based CBCT angiography system [26]. The last paper by Wan *et al.* discusses extraction of the hepatic vasculature in rats from a micro-CT image volume [38].

II. Overview and Future of the Field

With the rapid development of spiral/helical MDCT, spiral/helical CBCT should be the future of medical X-ray CT. The principal difference between MDCT and CBCT is that the former implies that the cone-angle is insignificant, whereas the latter takes the cone-angle into account so that more detector rows can be used. As in MDCT, the cone-angle is only a couple of degrees and can be ignored in image reconstruction. In the near future, the small cone-angle will likely be expanded to medium cone angles by adding more detector rows. Consequently, correction for the cone-beam effect should be included. Eventually, for large cone-angles, reconstruction algorithms must rigorously accomodate the cone-beam geometry. Furthermore, unlike other cone-beam CT systems, spiral/helical CBCT only collects *longitudinally truncated* projection data along a helical scanning locus. Over the past two decades, various CBCT algorithms have been developed. In the following, we share our observations on advantages/disadvantages of these algorithms.

A. Exact Algorithms

Traditional exact cone-beam reconstruction algorithms were designed according to the theory developed by Smith [31], Tuy [37], and Grangeat [14]. The sufficient condition for exact cone-beam reconstruction is that "*on every plane that intersects the object there exists at least one cone-beam source point*" [31]. When this condition is satisfied, the Grangeat formula relates the line integral of cone-beam data to the planar integrals in the 3-D radon space for exact image reconstruction. However, these exact algorithms assume that every cone-beam projection must completely contain an object to be reconstructed; hence, they cannot be used in spiral/helical CBCT, where projection data are longitudinally truncated.

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With a recent generalization of the Grangeat formula [20], [35], exact image reconstruction is possible in spiral/helical CBCT with longitudinally truncated data [6], [20], [21], [25], [35], [36]. The fundamental concept is to synthesize a mosaic of truncated fan-beams in any plane intersecting an ROI, apply the generalized Grangeat formula to triangular sections defined by the fan-beams, and accumulate their contributions to recover the planar integrals needed for exact reconstruction. A key to understanding the completeness of such data is to recognize that the detection window delineated by two consecutive turns of the helical path on the scanning cylinder allows a seamless X-ray coverage of the ROI [34].

B. Approximate Algorithms

Feldkamp *et al.* [8] heuristically adapted the fan-beam algorithm for approximate cone-beam reconstruction in the case of a circular locus. The Feldkamp algorithm was generalized for spiral/helical CBCT [39]–[41], [45] to allow either full-scan or half-scan data from flexible 3-D loci, including a helix. In addition to these Feldkamp algorithms, other approximate algorithms were recently developed in the filtered-backprojection framework [3], [4], [15], [16], [19], [24], [27], [29], [30], [46].

Despite elegant results on exact cone-beam reconstruction, approximate cone-beam algorithms remain practically and theoretically valuable. In the ideal case that projection data are complete, consistent, noise-free, and manipulated in the continuous domain, the exact approach would be the method of choice. However, measurement noise, missing data, data interpolation, patient motion, digital processing, and dose utilization must be taken into account in practice, which have significant impacts on image quality. Relative to the exact spiral/helical CBCT algorithms, the approximate image reconstruction algorithms have the following advantages.

First, the approximate algorithms may produce better spatial/contrast resolution. An important fact is that projection data are necessarily contaminated by noise, especially in low-dose CT screening and CT fluoroscopy. Because the approximate reconstruction requires only one-dimensional (1-D) data filtering, instead of two-dimensional (2-D) data filtering (differentiation is considered as equivalent to filtration) required in the exact reconstruction, the approximate reconstruction needs less regularization than does the exact reconstruction for a given image noise level. In other words, the spatial resolution is less compromised in the approximate reconstruction in this situation. It is possible to tolerate higher noise levels and produce a better spatial resolution, but the contrast resolution would be reduced. For example, exact cone-beam reconstruction with the direct Fourier method produces more noise and ringing compared with the Feldkamp method [1]. The problems may be inherent to all exact cone-beam reconstruction algorithms that filter data two-dimensionally. Furthermore, data interpolation is less involved in the approximate algorithms; hence, less image blurring will occur from this type of data processing, relative to "exactly" reconstructed images.

Second, the approximate algorithms produce better temporal resolution. In the exact cone-beam reconstruction algorithms, multiple source points are needed to define a partition of each plane through an ROI. Therefore, the angular range of projection data may span several helical turns. In contrast to the extensive data range of the exact algorithms, the approximate algorithms work only with either full-scan or half-scan data. The less the angular range involved, the better the temporal resolution, which is critical for cardiac imaging, lung imaging, and CT-guided medical interventions.

Third, the approximate algorithms are computationally more efficient. The approximate reconstruction involves less raw data; hence, it requires less computing resources. The computational structure of the approximate reconstruction is relatively straightforward, highly parallel, and hardware supportable. The approximate algorithms are particularly faster for reconstruction of a limited number of slices or small ROIs.

The power of the approximate algorithms can be further appreciated from the relationship between the exact reconstruction and the Feldkamp reconstruction. On the one hand, the exact reconstruction naturally degrades to the Feldkamp reconstruction when the scanning locus is circular [5], [31]. On the other hand, there is a fundamental link from the Feldkamp reconstruction to the exact reconstruction [44]. Briefly speaking, according to the steps of the Feldkamp reconstruction, the approximate reconstruction may be achieved in a rotated reconstruction system. Being consistent to the well-known property of the Feldkamp algorithm that the vertical integral of a reconstructed volume is exact, it can be proved that the integral of a reconstructed image along the rotated z-axis is also exact. In other words, exact 2-D parallel projections can be synthesized along various directions. If the sufficient condition for exact cone-beam reconstruction is satisfied, a sufficient amount of exact 2-D parallel projections can be synthesized using the Feldkamp approach; hence, the exact image reconstruction can be performed.

C. Iterative Algorithms

In addition to the exact and approximate spiral/helical CBCT algorithms, which are based on closed-form inversion of projection data, iterative algorithms may be important in medical X-ray CT because of their capability of handling photon fluctuation and the ever-improving computing technology. A major common weakness of the noniterative algorithms, either exact or approximate, is that projection data are implicitly assumed to be noise-free. However, noise is an inherent aspect of projection data, especially for low-dose scans. Because CT can be viewed as a parameter estimation problem in the case of low-count projection data, the statistical approach should have a central role in image reconstruction [11].

Iterative statistical reconstruction methods have been successful in emission CT, primarily in PET and SPECT [25]. Since the publication of the first iterative X-ray CT algorithm for the maximum likelihood (ML) reconstruction [22], several new iterative X-ray CT algorithms [23], [9], [10] were proposed to accelerate the original ML algorithm. The recent algorithm based on paraboloid surrogates [7] is guaranteed to be monotonic with the computing time comparable to that of penalized weighted least squares (PWLS) methods [11].

The expectation maximization (EM) algorithm for emission CT was adapted for X-ray CT according to the I-divergence minimization interpretation [32], [42]. It was demonstrated that the EM emission CT algorithm, also referred to as the iterative deblurring algorithm, is promising for image noise suppression and metal artifact reduction in X-ray CT. Furthermore, the EM emission CT algorithm can be greatly accelerated using the ordered-subset or row-action scheme [2], and may be adapted for X-ray CT, especially in the case of CT fluoroscopy [43].

Iterative algorithms have potential with further development to surpass the exact or approximate algorithms in important applications, including image denoising, metal artifact reduction, local region reconstruction, and CT fluoroscopy. Iterative algorithms can be either approximate or exact (in the sense of the limiting case), depending on the completeness of the scanning and detection geometry. Parallel/array processing techniques are particularly valuable for implementation of iterative algorithms.

III. Concluding Remarks

All three types of spiral/helical CBCT algorithms have unique advantages. The exact algorithms can produce accurate images when data are complete and noiseless. The approximate algorithms may outperform the exact algorithms when data are noisy or involved with moving structures. The iterative algorithms should be preferred when data are very noisy or incomplete. Approximate and iterative algorithms may produce artifacts when data are sparse. In addition to our qualitative comments on these spiral/helical CBCT algorithms, quantitative comparison should be performed for specific tasks. Given a generic CT algorithm, image quality indexes are not uniquely specified. For example, spatial resolution and contrast resolution can be altered by either preprocessing data or postprocessing images. The optimal protocol for any application should be established according to clinical requirements, based on comparative studies, and using relationships among image quality indexes and imaging parameters.

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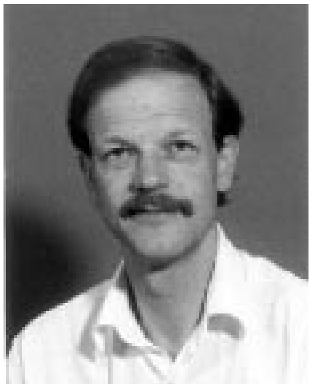

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