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Cone-beam digital tomosynthesis for thin slab objects

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ABSTRACT

We describe a cone-beam computed tomography with insufficient projections obtained from a limited angle scan, the so-called cone-beam digital tomosynthesis. Digital tomosynthesis produces cross-sectional images parallel to the axis of rotation from a series of projection images acquired from a planar detector. The image reconstruction algorithm is based on the cone-beam filtered backprojection method. To suppress the out-of-plane artifacts due to the incomplete sampling over a limited angular range, we applied an apodizing filter in the depth direction. We applied the digital tomosynthesis technique to a multilayer printed circuit board possessing thin slab geometry and evaluated its performance with respect to various operation parameters, such as the total scan angle, the step angle and the number of projection images used for reconstruction. The results showed that the image quality of digital tomosynthesis reconstructed for the total scan angle greater than 60 degrees with a step angle as narrow as possible exhibited that it was comparable to that of the computed tomography. The digital tomosynthesis technique is expected to be practical for extracting internal cross-sectional views, parallel to the scan direction, of objects with thin slab geometry.

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1. Introduction

Although x-ray computed tomography (CT) was at first born and now is widely used in medical diagnoses [1,2], it has also been essential in industrial quality control and nondestructive evaluation [3–5]. The recent cone-beam CT (CBCT) with large-area flat-panel detectors, enabling three-dimensional (3D) volume imaging in a single scan, pushes its necessity further forward both in medicine and industry.

To achieve an artifact-free tomography, CT requires as many projections as possible over a single circular scanning motion around an object. Although a rigorous determination of the minimum number of projections is less straightforward [6], the number of projections per single circular scan in fan-beam geometry is typically of the order of the number of line-detector elements. However, many situations are restrictive in obtaining the adequate number of projections; the size as well as the geometrical complexity of an object to be imaged might sometimes restrict the complete traverse motion of the CT gantry system. Even though the complete scan is available, the CT for an object having thin slab geometry, for example plate-like printed circuit boards (PCBs), is impractical [7] because the conventional CT typically defines the reconstruction slices perpendicular to the axis of rotation (AOR) [8]. Improvement in the density of PCBs and ball grid array (BGA) packages highly demands cross-sectional images parallel to the AOR [9], and the x-ray laminography has been a popular tomographic technique that produces an interesting planar cross-sectional view using relative motions of the x-ray source, object and detector [7,10,11].

For the scanning in certain limited angular ranges, the limited angle tomography is now regarded as a new genre in medical and industrial imaging techniques, popularly known as digital tomosynthesis [12]. Tomosynthesis is a technique that produces section, or slice, images parallel to the AOR from a series of projection images acquired as the x-ray source moves over a prescribed path. Therefore, it may provide 3D information at a lower x-ray irradiation and potentially lower cost than the conventional CT in certain imaging situations. It was also reported that the spatial resolution in tomosynthesis images is better than that of a CT by a factor of three [13]. Although the theoretical framework describing tomosynthesis had been initialized in the 1930s, its renewed interest has been in the spotlight since large-area flat-panel detectors had been commercially available in the late 1990s [14].

In this paper, we describe the theoretical framework of a modified filtered-backprojection (FBP) method in digital tomosynthesis and demonstrate its feasibility in the application to a multilayer PCB.

2. Theoretical framework

The basic reconstruction algorithm for tomosynthesis is the shift-and-add (SAA) method that shifts each of the projection

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images by a given amount and then adds them together to make objects in a given plane to be sharpened while objects from other planes to be blurred [15]. This SAA method is basically equivalent to the simple backprojection method [16]. It is, however, insufficient to use SAA alone for high-quality tomosynthesis because of blur from outside of the plane of interest (POI). Therefore, the SAA or backprojection method requires deblurring procedures to compensate for the out-of-plane blur [17].

We adopt the modified FBP method for the cone-beam digital tomosynthesis (CBDT) with insufficient projection data obtained from a limited angle scan. The strategy is based on the conventional FBP method with additional filters to control the incomplete frequency responses due to sampling in a limited angular range in cone-beam geometry [18].

2.1. Reconstruction algorithm

Cone-beam geometry with planar detectors is very attractive because of the simple geometry and quick scan for large volumetric objects, and thus it is popular in many technical applications. However, since a single circular x-ray source trajectory in cone-beam geometry does not provide complete Radon data, the exact 3D image reconstruction is not available following the Tuy–Smith sufficiency condition [19]. Instead, an approximate version of the exact reconstruction developed by Feldkamp, Davis and Kress, the so-called FDK method [20], is typically used. The FDK method is a simple extended version of the conventional FBP method in longitudinal direction by considering the cone angle [3]. It is, therefore, an approximate approach and may cause distortion when the cone angle is wide. However, since the FDK algorithm is very fast and easily applicable, it is most widely used in commercial CBCT systems [3].

Fig. 1 illustrates the reconstruction scheme of a voxel located at a plane (x, z) at a depth y, denoted by $f(x, z; y)$, in an object with the projection data $p(\xi, \eta; \beta)$ obtained at a projection angle β. Following the notations described in Fig. 1, the CBDT algorithm can be given by [21]

$$f(x,z;y)=\int_{\beta_{min}}^{\beta_{max}}\frac{d^2}{(d-s)^2}\int_{-\infty}^{\infty}\frac{L}{\sqrt{L^2+\xi^2+\eta^2}}p(\xi,\eta;\beta)h\left(\frac{L}{d-s}t-\xi\right)d\xi\,d\beta, \tag{1}$$

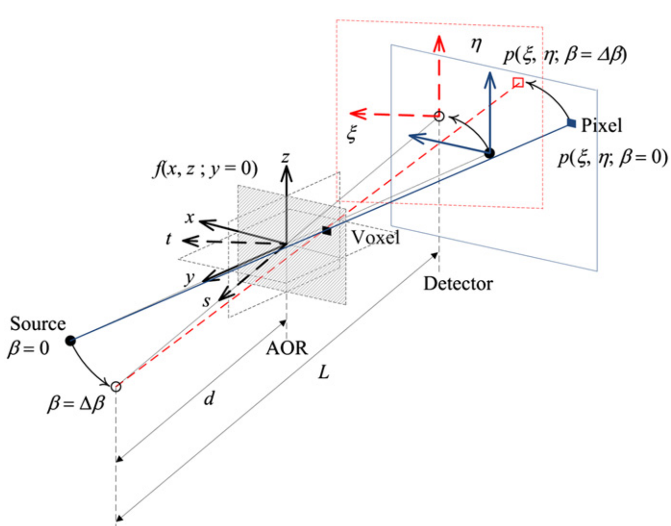

Fig. 1. Sketch describing image reconstruction in cone-beam geometry. For the reconstruction of a voxel located at a plane (x, z) at a depth y in an object, the contribution of projection values at (ξ, η) in the planar detector obtained at two different projection angles, $\beta=0$ and $\Delta\beta$, is illustrated. The projection signal is backprojected along the line which passes the source and the pixel value in the detector.

where L refers to the source-to-detector distance and d the source-to-AOR distance. The (t, s)-system denotes the rotated (x, y)-system, and ξ and η are the detector axes perpendicular and parallel to the AOR, respectively. The sequential calculation steps are as follows: (i) the projection data are weighted by the term $L/\sqrt{L^2+\xi^2+\eta^2}$, known as the cosine weighting; (ii) the pre-weighted projection data are convolved with a one-dimensional (1D) ramp filter $h(\cdot)$ to suppress low-frequency information in the projection data and (iii) the filtered projection data are distance-weighted by the term $d^2/(d-s)^2$, and then backprojected to the POI. Therefore, Eq. (1) describes that the 1D filtered pre-weighted projection signals are added in each virtual voxel considering magnification between the voxel and the corresponding projected pixel over the scanned angular range. CBDT is similar to the FDK-type CBCT solved only for specified depths y with a total scan angle $\beta_{scan}=\beta_{max}-\beta_{min}$.

2.2. Additional filter design for tomosynthesis

If we designate u, v and w as the Fourier conjugates corresponding to x, y and z, respectively, in the space Cartesian coordinates, the high-pass or ramp filter in Eq. (1) can be described in the Fourier domain as [22]

$$H_{RA}(u,w)=\beta_{scan}\sqrt{u^2+w^2}, \tag{2}$$

where β_{scan} denotes the total scan angle. Since this ramp filter is vulnerable to high-frequency noise, we combined it with a spectral apodizing filter:

$$H_{SA}(u)=\frac{1}{2}\left[1+\cos\left(\frac{\pi u}{k_{SA}}\right)\right], \tag{3}$$

which limits the bandwidth of the projection data, hence reducing high-frequency noise, spectral leakage and aliasing. The parameter k_{SA} determines the bandwidth and is typically multiples of the Nyquist frequency in u direction, u_N.

Incomplete sampling over the limited angular range results in an out-of-plane artifact in the reconstructed images [23]. We incorporated an additional slice thickness filter to limit the frequency response in the depth or y direction as

$$H_{ST}(v)=\frac{1}{2}\left[1+\cos\left(\frac{\pi v}{k_{ST}}\right)\right]. \tag{4}$$

The slice thickness filter $H_{ST}(v)$ plays the same role as the spectral apodizing filter but the application direction is different. Similarly, the parameter k_{ST} determines the bandwidth and is typically multiples of $u_N\times\tan(\beta_{scan}/2)$ [24]. This filtering operation results in "anisotropic" voxel size to avoid aliasing artifact, consequently a scan-angle-dependent slice thickness which is inversely proportional to $\tan(\beta_{scan}/2)$. It should be noted that we realized the two apodizing filters with the Hann window functions [25].

3. Materials and methods

3.1. Experimental

Instead of gantry rotation, we employed an object-rotation system to acquire projection images for CBDT. The imaging detector and the x-ray tube were fixed with a distance of 500 mm in an upright position. A rotational object jig was attached at a linear guide bar, which had linear translational motion against the x-ray tube located at the bottom side of the imaging configuration.

X-ray image intensifier (XRII) system (TH 9464, Thales Electron Device, France) was used as the imaging detector. The XRII

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