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ABSTRACT: We present a volumetric dental tomography method that compensates for insufficient projection views obtained from limited-angle scans. The reconstruction algorithm is based on the backprojection filtering method which employs apodizing filters that reduce out-of-plane blur artifacts and suppress high-frequency noise. In order to accomplish this volumetric imaging two volume-reconstructed datasets are synthesized. These individual datasets provide two different limited-angle scans performed at orthogonal angles. The obtained reconstructed images, using less than 15% of the number of projection views needed for a full skull phantom scan, demonstrate the potential use of the proposed method in dental imaging applications. This method enables a much smaller radiation dose for the patient compared to conventional dental tomography.

KEYWORDS: Computerized Tomography (CT) and Computed Radiography (CR); Medical-image reconstruction methods and algorithms, computer-aided so; X-ray radiography and digital radiography (DR)
1 Introduction

Computed tomography (CT) has become one of the most frequently used imaging modalities used for the preoperative evaluation of the jaw for dental implants. As a replacement for a conventional diagnostic CT system, a dedicated dental CT system has recently been developed [1]. Although dental CTs typically require an imaging detector with a field of view (FOV) greater than $100 \times 100$ mm$^2$ to cover both the maxilla and mandible, a smaller FOV covering just a few teeth is sometimes sufficient for an image-guided implant operation, which can benefit local dentists who need a cost-effective dental CT system. However, the actual CT realization using the truncated projection obtained from a small-FOV detector is problematic. The digital tomosynthesis technique, which can produce sectional images parallel to the axis of rotation from a series of projection views acquired as the x-ray source moves over a prescribed path, could be used to mitigate this problem [2]. However, digital tomosynthesis only provides slice images parallel to the scan direction due to its poor depth resolution [3], thus making volume imaging with digital tomosynthesis impractical.

In this study, we present a digital tomosynthesis technique that enables volumetric dental imaging. The proposed approach synthesizes two volume-reconstructed datasets, taken separately, for two different limited-angle scans performed at orthogonal angles, thereby enhancing the depth resolution.

2 Methods and materials

As shown in figure 1(a), the proposed dental tomosynthesis employs an isocentric linear motion similar to a conventional single-circular cone-beam CT motion; however the scan trajectories are two at orthogonal angles, such as longitudinal and cross-sectional. The theoretical framework for the image reconstruction is based on the work of Lauritsch and Härer who discussed a filtered backprojection method used for circular tomosynthesis [4]. The principle is similar; however we employ a backprojection filtering method that is more appropriate for incomplete scan trajectories.

In order to accomplish digital tomosynthesis for isocentric linear motion, we made some assumptions in regards to parallel-beam geometry, continuous sampling, and the linear and shift-invariant (LSI) process in the projection-backprojection operation [4, 5]. From the LSI approximation, a three dimensionally (3D) reconstructed image $g(x,y,z)$ can be described by the convolution
of an object function $f(x, y, z)$ with a 3D point-spread function $h(x, y, z)$ by using a projection-backprojection operation. Utilizing forward and inverse Fourier transformations as well as simple arithmetic manipulation, we can restore the object function by:

$$f(x, y, z) = \mathbf{F}^{-1}_3 \left\{ H^{-1}(\omega_x, \omega_y, \omega_z) \cdot G(\omega_x, \omega_y, \omega_z) \right\},$$

(2.1)

where all the capital letters denote the respective Fourier conjugates to the functions defined in the space domain and $\omega_j$ is the spatial frequency corresponding to the $j$ direction in the space Cartesian coordinates. $\mathbf{F}^{-1}_3$ denotes the 3D inverse Fourier transformation. The determination of the filter function $H^{-1}(\omega_x, \omega_y, \omega_z)$ is the key procedure needed in the image reconstruction.

In conventional CT, the filter function uses a well-known ramp filter. This ramp filter is vulnerable to high-frequency noise which can be suppressed by an appropriate apodizing filter, i.e. a band-limited window function that removes or smooths the discontinuities at the beginning and end of the sampled data. We note that there are additional discontinuities, for example, at $|\omega_z| = |\omega_x| \tan(\alpha/2)$, in which $\alpha$ is the scan angle, for the sampled data from the longitudinal scan, as shown in figure 1(b); these may cause ringing artifacts in the reconstructed images and increase the blur artifacts originating out of plane [4]. These behaviors can also be suppressed in a similar manner, which ensures a constant depth resolution over a wide range of spatial frequencies. Therefore, for limited angle tomography employing isocentric linear motion and parallel-beam approximation, we can employ:

$$H^{-1}(\omega_x, \omega_y, \omega_z) = A(\omega_x, \omega_y) \cdot B(\omega_x, \omega_z) \cdot C(\omega_x, \omega_z),$$

(2.2)

where $A(\omega_x, \omega_y)$ is the angular weighted ramp filter ($=2\alpha \sqrt{\omega_x^2 + \omega_y^2}$) [5]. $B(\omega_x, \omega_z)$ is the apodizing filter used to control the high-frequency noise in the ramp filter. $C(\omega_x, \omega_z)$ is the additional

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**Figure 1.** An illustration of the Fourier-slice theorem and the incompleteness of the object information in limited angle tomography. The projection dataset in the space domain (a) is mapped into the Fourier domain (b) in a double-wedge shape.
apodizing filter used to suppress the frequency response of the out-of-plane blurred structures. For the implementation of both apodizing filters, we used the Hann function known as a raised cosine window [6].

Since the single or longitudinal scan shown in figure 1(a) gives poor depth resolution [3], the cross-sectional images extracted parallel to the depth direction typically show poor image quality. In order to overcome this problem, we present a method that synthesizes two individual reconstructed-volume datasets for two different limited-angle scans, i.e. longitudinal and cross-sectional scans, as described in figure 1.

In order to verify the proposed volumetric dental imaging approach, we applied the algorithm to a commercial dedicated dental CT system (Implagrapy™, Vatech Co., Korea). As shown in figure 2(a), the dental CT system employed a cone-beam geometry and used an indirect-detection flat-panel x-ray detector with an FOV of $80 \times 50 \text{ mm}^2$. The pixel pitch of the detector was $200 \mu\text{m}$. The CT scan was performed upon a skull phantom (Model PUT-2, Kyotokagaku, Japan). The projection views were obtained using an 85-kVp W x-ray spectrum with the beam current of 7 mA. In order to mimic a limited angle geometry, we took 53 projections for each $52^\circ$ scan angle amongst the full scan dataset. In other words, emulating figure 1(a), the longitudinal scan was performed for the angular range of $244^\circ$ to $296^\circ$ and the cross-sectional scan of $344^\circ$ (or $-26^\circ$) to $26^\circ$. We note that the dental CT system generates 720-projection views for a full $360^\circ$ scan in 24 seconds.

For a comparison, the image reconstruction based on the conventional shift-and-add method, the traditional algorithm in digital tomosynthesis [7], was performed. The shift-and-add method shifts each of the projection views by a given amount and then adds them together to make objects in a given plane sharper while blurring the objects in other planes.

As designated in figure 2(b), whereas the longitudinal scan provides a tomogram parallel to the $A-A'$ directional plane, the cross-sectional scan provides a tomogram parallel to the $B-B'$ plane. In order to prove the volume imaging capability of the proposed method, we extracted a tomogram parallel to the $C-C'$ plane from the synthesized volume-reconstructed data.

3 Results

The tomosynthesis image, parallel to the $A-A'$ directional plane, as designated in figure 2(b), obtained for the longitudinal scan using the dental CT system, is shown in figure 3(a). The image size is $256 \times 256 \times 1$ voxels. For the comparison the tomogram obtained by using the shift-and-add method is shown in figure 3(b). As expected, the image reconstructed by using the shift-and-add method is blurry because the method used is basically equivalent to the simple backprojection method [7]. Figure 3(c) shows the tomosynthesis image, parallel to the $B-B'$ directional plane designated in figure 2(b), obtained for the longitudinal scan. Due to the lack of sampled data in the depth direction, as shown in figure 1(b), the quality of the tomogram is very poor, making it difficult to identify any anatomical structures.

Figure 4(a) shows the volume-rendered image synthesized after the additional cross-sectional scanning and image reconstruction procedures. The total number of projection views was doubled to 106 projection views, however, that is only 14.7% of the full 720-projection CT scan data. The tomosynthesis image parallel to the $B-B'$ directional plane, as designated in figure 2(b), is illus-
Figure 2. (a) The dedicated dental CT system used for the digital tomosynthesis experiment and (b) the reconstructed plane directions for molar teeth in the mandible. The A–A’ and B–B’ directional planes are used for the longitudinal and cross-sectional scans, respectively. The C–C’ plane is extracted from the synthesized volume-reconstructed data.

Figure 3. Illustrative tomosynthesis images obtained for the longitudinal scan. (a) The tomogram parallel to the A–A’ directional plane, as designated in figure 2(b), obtained by the digital tomosynthesis technique for the skull phantom. For the comparison, the tomogram obtained by using the shift-and-add method (b) is shown. (c) A cross-sectional image perpendicular to (a) [or parallel to the B–B’ directional plane] which shows poor image quality.

Figure 4. Illustrative images obtained for the cross-sectional scan. (a) The tomosynthesis image perpendicular to the A–A’ directional plane, as designated in figure 2(b), obtained by the digital tomosynthesis technique for the skull phantom. For the comparison, the tomogram obtained by using the shift-and-add method (b) is shown. (c) A cross-sectional image perpendicular to (a) [or parallel to the B–B’ directional plane] which shows poor image quality.
Figure 4. (a) The volume-rendered image synthesized after two perpendicular directional scans and image reconstruction procedures. (b) The tomogram parallel to the B–B' directional plane as designated in figure 2(b). The arrow indicates the mandibular canal. (c) The tomogram obtained at an arbitrary angle direction parallel to the C–C' directional plane as designated in figure 2(b) used to enhance the visualization of the mandibular canal designated by the arrow.

4 Discussion and conclusion

Instead of the analytic approach described in this study, an iterative image-reconstruction approach could be used for limited angle tomography. However, it has been reported that the iterative approach, as in the expectation maximization method for example, also suffers from the sparse-view artifacts due to its data sampling incompleteness [7, 8].

Since the introduction of the compressed sensing theory, which enables nearly ideal images to be reconstructed even if there are only a few projections [9], a great deal of effort has been devoted to image reconstruction from data collected from significantly fewer projection views than what is used in current cone-beam CT methods based on flat-panel detectors, although the outcomes are still to be determined [8]. However, a practical implementation of this approach still remains a challenge because of the iterative nature involved in solving compressed sensing formulations, hence the expensive computational cost.

The issues raised by the expensive computational cost needed for image reconstruction algorithms incorporating intensive iterative procedures could be remedied by a computationally efficient data-parallelization approach with proper hardware, graphics processing units for example, and/or mathematically formulating an efficient search algorithm that provides a fast solution convergence [10]. Although it is obvious that this advanced image-reconstruction method will replace the conventional simple methods due to its reduction in the patient’s radiation dose as well as a better image quality, it is not expected to be implemented in clinics at present.

We have presented a backprojection filtering method for limited angle scans and demonstrated its usefulness in dental imaging. The number of required projection views is much smaller than that the number typically used for conventional dental CT scans and the quality is promising. The digital tomosynthesis for the dental imaging described in this study has the potential to be used for planning an implant procedure with a greatly reduced patient dose without significantly altering the operation protocol requiring a conventional dental CT system. Compared to the image quality
obtained by a full scan, the optimization based on the quantitative evaluations of image quality with respect to various imaging parameters, such as the number of projection views and the span of scan angle, is left for the future studies.

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References