Virtual Colonoscopy Screening with Ultra Low-Dose CT: A simulation study

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Abstract --- **Computed tomography colonography (CTC) or CT-based virtual colonoscopy (VC) is an emerging tool for detection of colonic polyps. Compared to the conventional fiber-optic colonoscopy, VC has demonstrated the potential to become a mass screening modality in terms of safety, cost, and patient compliance. However, current CTC delivers excessive X-ray radiation to the patient during data acquisition. The radiation dose is a major concern for screening application of CTC. In this work, we performed a simulation study to demonstrate a possible ultra low-dose CT technique for VC. The ultra low-dose abdominal CT images were simulated by adding noise to the sinograms of the patient CTC images acquired with normal dose scans at 100 mAs levels. The simulated noisy sinogram or projection data were first processed by a Karhunen-Loève domain penalized weighted least squares (KL-PWLS) restoration method and then reconstructed by a filtered backprojection algorithm for the ultra low-dose CT images. The patient-specific virtual colon lumen was constructed and navigated by a VC system after electronic colon cleansing of the orally-tagged residue stool and fluid. By the KL-PWLS noise reduction, the colon lumen can be successfully constructed and the colonic polyp can be detected in an ultra low-dose level below 50 mAs. Polyp detection was also found easier by the KL-PWLS noise reduction compared to the results using the conventional noise filters, such as Hanning filter. These promising results indicate the feasibility of an ultra low-dose CTC pipeline for colon screening.**

I. Introduction

COLONIC polyps have a high probability (greater than 90%) of developing into colorectal cancer, which is currently the second leading cause of deaths related to cancers. Early detection and removal of colonic polyps can significantly reduce the risk of death. The American Cancer Society has recommended a colon examination every three to five years for people of age over 50. Fiber-optic colonoscopy (OC) is currently the most commonly used diagnostic procedure. However, patients are usually reluctant to take the OC procedure because it is invasive, time consuming, and expensive. Computed tomography colonography (CTC) or CT-based virtual colonoscopy (VC) is an emerging tool for colon polyp detection [1][2]. Although VC is minimal- or non-invasive and less stressful to the patient, CTC is not totally risk-free. The radiation in forms of X-ray exposure to the patient during CT scan could lead to adverse health effects in a later time of the patient's life. Minimizing the radiation risk to the patient, while maintaining satisfactory CT image quality, becomes urgent for colon screening with CTC.

Dose reduction for CT imaging can be achieved by acquiring projection data with low-mAs protocols. With a low-mAs acquisition protocol, noise due to less X-ray photons will degrade the CT image quality. In the past years, research effort on both hardware optimization and noise filtering on acquired data has made noticeable progress from normal dose scans (over 100 mAs levels) down to low-dose CTC scans as low as 50 mAs level [2]. Further decreasing the mAs level for ultra low-dose CTC scans may induce streak artifacts in the reconstructed images if oral contrast solutions are used to tag the stool and colonic fluid because the contrast solutions absorb a noticeable amount of X-rays [3]. Therefore, more sophisticated noise treatment than a simple low-pass noise filtering is necessary. Recently a statistics-based framework of sinogram restoration followed by filtered backprojection (FBP) image reconstruction has shown promising results for ultra low-dose CT [4]-[9]. This framework is based on noise modeling of the projection data, where the noise modeling is similar to that of a statistical iterative image reconstruction approach. Difference between this sinogram restoration framework and the iterative image reconstruction approach is that the penalty or the smoothing constraint in the sinogram restoration framework is in the sinogram space while the penalty for the iterative image reconstruction approach is in the image domain. In our previous work [6], we have shown a similar performance between a statistics-based sinogram

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restoration strategy and a statistical iterative reconstruction algorithm in terms of image quality and detectability in low-contrast environment, where both methods seek the same solution of minimizing the penalized weighted least-squares (PWLS) cost function of the data distribution which was simulated from anthropomorphic digital phantoms. However, the iterative image reconstruction algorithm consumed a great computing power for volumetric CT and might not be practical for clinical use. For example, the reconstruction time for a routine clinical study consisting of several hundred slice images of 512×512 array size is at the order of hours by a currently available fastest PC platform. The sinogram restoration strategy is much more efficient (with more than ten fold reduction of computing time) and has the potential to be utilized in real-time clinical situations.

In this work, we performed a simulation study using patient CTC images to demonstrate a possible ultra low-dose CT technique for VC screening purpose. The ultra low-dose projection data were simulated from the patient volumetric CTC images based on the noise properties of clinical CT projection data [5][10][11] and had a noise level less than 50 mAs of a clinical low-dose scanning protocol. The simulated ultra low-dose projection data were first processed by the statistics-based Karhunen-Loève domain PWLS (KL-PWLS) restoration strategy [4][6] and then reconstructed by a standard FBP algorithm (i.e., by the use of the Ramp filter at 100% Nyquist frequency cutoff). Promising results were obtained, demonstrating the feasibility of an ultra low-dose CTC pipeline for colon screening.

II. Methods

A. Simulation of Low-Dose CT Sinograms

An ultra low-dose CT scan of the abdomen was simulated from a corresponding patient CTC volume image acquired at a routine normal dose level, i.e., acquired by a protocol with a mAs value around 100 (or 200 mA at a rotation speed of 0.5 seconds per rotation). The sinograms (or line integrals) of the patient abdominal CTC volume image were calculated in a slice-by-slice fashion. The simulated sinograms from all the image slices mimic the experimental sinogram data in Radon space after system calibration, which includes interpolation of spiral-sampled projection data from multi-detector bands, data conditioning via the logarithm transformation, uniformity calibration on detector cells' responses, etc. The geometry used to generate the sinograms is based on a commercial (e.g., GE) CT scanner. The number of detector cells per view is 888. A total of 984 views spans evenly on a circular orbit of 360^{o}. The detector arrays in each band are on an arc concentric to the X-ray source with a distance of 949.075 mm. The distance from the rotation center to the X-ray source is 541 mm. The detector cell spacing is 1.0239 mm.

Each of the 888 line integrals at each of the 984 view angles was calculated based on the Siddon's ray-tracing technique [12] between the X-ray source point and the center of the detector cell. The intersecting length of the ray with a square image pixel was used as the weight of the pixel's contribution to the line integral. The calculated sinograms of the fan-beam geometry contained a noise level equivalent to a normal dose level of the original CTC scan. Lower dose scans can be simulated by adding corresponding noise levels [15]. It has been shown a lower dose scan can be simulated by adding Gaussian noise with a variance dependent on the mAs level of the normal scan [13]-[15]. In our previous experimental studies [5][10][11], the noise in CT sinogram (line integrals after logarithmic transform) was shown to have a signal-dependent variance. The variance of projection datum or line integral $\bar{p}_i$ at detector cell *i*, $\sigma^2_{p_i}$, can be estimated by

$$\sigma^2_{p_i} = f_i \exp(\bar{p}_i / \eta) \quad (1)$$

where η is a scanner-specific scaling parameter and f_i represents an adjustable factor adaptive to each detector cell across the field-of-view (FOV) and considers mainly the effect of different incident photon numbers at different detector cells. Given a sinogram acquired at a high mAs level, a lower mAs sinogram can be simulated by adding Gaussian noise with a corresponding curve of $\{ f_i \}$ across the FOV.

In this study, the patient CTC images were acquired at high mAs levels (or normal dose scans). The computed line integrals of each patient CTC image reflect a corresponding high mAs level. By selecting a suitable curve $\{ f_i \}$ [11] across the 888 detector cells or at each projection view, a low mAs sinogram was then simulated. We simulated ultra low-dose sinogram for each patient at mAs levels below 50. The mAs can be as low as 10 mAs, depending on the patient size and the amount of oral contrast solution used to tag the colonic materials.

B. Statistics-Based Sinogram Restoration for Ultra Low-Dose CT

The simulated ultra low-dose (or noisy) three-dimensional (3D) sinogram was processed by the KL-PWLS restoration strategy [4][6]. The KL-PWLS strategy is a statistics-based algorithm that aims to estimate the ideal sinogram by minimizing the PWLS objective function in the KL domain,

$$\Phi_l(\tilde{q}_l) = (\tilde{y}_l - \tilde{q}_l)' \tilde{\Sigma}_l^{-1} (\tilde{y}_l - \tilde{q}_l) + (\beta / d_l) \tilde{R}(\tilde{q}_l), \quad (2)$$

where $\tilde{y}_l$ and $\tilde{q}_l$ are the *l*-th KL principal components of the noisy sinogram p and the ideal sinogram $\bar{p}$ (to be estimated) respectively, and $\tilde{\Sigma}_l$ is the diagonal variance matrix of $\tilde{y}_l$. In the KL domain, each KL component is a 2D image array. Notation d_l indicates the eigenvalue of the *l*-th KL component, β is a smoothing parameter which controls the degree of agreement between the estimated and the

measured data via the penalty term $\widetilde{R}(\widetilde{q}_l)$ which usually has a quadratic form of

$$\widetilde{R}(\widetilde{q}_l) = \widetilde{q}_l' \widetilde{R} \widetilde{q}_l = \tfrac{1}{2} \sum_i \sum_{m \in N_i} w_{im} (\widetilde{q}_{i,l} - \widetilde{q}_{m,l})^2 \qquad (3)$$

where N_i indicates the nearest or first-order neighbors of the *i*-th pixel in each KL component along the bin and view directions (after the KL transform was applied along the direction of rotation) and parameter w_{im} is equal to 1 for the first-order neighbors.

This objective function models the first and second moments of the sinogram data, where the non-stationary noise property is accurately considered by equation (1) which specifies the weights or the diagonal elements in matrix $\widetilde{\Sigma}_l$ in the PWLS criterion. The KL transform models the correlation among neighboring views of the sinogram data and, therefore, provides a data-adaptive penalty role. The goal of minimizing the objective function is to find an optimal solution based on the data statistics and data correlation. After the KL transform, the chosen neighboring views of sinogram were decomposed to several independent KL principal components. Each KL component is associated with a KL eigenvalue, which reflects a corresponding signal-to-noise ratio (SNR) of the KL component. A larger KL eigenvalue is corresponding to a higher SNR, and this information provides a mechanism to control the smoothing strength at different KL components via the penalty. By setting the smoothing parameter inversely proportional to the eigenvalue at each KL component, i.e., β / d_l the KL component with lower SNR (small eigenvalue) will be smoothed more during the PWLS restoration [4][6][9].

The restored 3D sinogram was reconstructed by a standard FBP algorithm slice-by-slice in fan-beam geometry. In order to avoid the non-uniform noise propagation problem in fan-beam geometry, the intersecting area of fan-beam strip and square image pixel was used as the weight, rather than a bi-linear interpolation, in the backprojection step in the FBP algorithm [16].

C. Electronic Colon Cleansing by Partial Volume Image Segmentation

The reconstructed ultra low-dose CT images contained tagged residue stool and fluid inside the colon lumen. The tagging was carried out by ingesting oral contrast solutions during a period of one or two days prior to the CT scan and is necessary in order to differentiate the stool and colonic fluid from the colon wall. The tagged colonic materials were virtually removed from the CT images by an electronic colon cleansing (ECC) technique [17], which is based on a partial volume (PV) segmentation algorithm [18]. The PV image segmentation algorithm determines the tissue mixture percentages inside each voxel and therefore considers accurately the PV effect upon the colon wall due to the enhanced image density of the tagged materials. The segmentation algorithm models the image data statistics and seeks the maximum *a posterior* (MAP) solution [17][18]. The MAP solution was computed by the expectation-maximization (EM) algorithm [19]. The output of the ECC was a PV layer which covers the colon mucosa and reflects very useful clinical information on the mucosa. Within the PV layer enclosure, the colon lumen was cleansed by a region-growing strategy. The MAP-EM PV image segmentation-based ECC technique has shown advantages in improving the detection of colonic polyps [18].

D. Construction of Colon Lumen Models for VC

The cleansed colon lumen was fed into the V3D-Colon Module developed by Viatronix Inc. (Stony Brook, NY), where the virtual colon model of the patient was constructed. The V3D-Colon Module simulates the navigation procedure of the clinical OC and provides a volume-rendered 3D endoscopic view at each location on a centerline of the virtual colon model. During the fly-through navigation along the centerline, the user has the control on the navigation speed and the view angle to facilitate colon polyp detection [1]-[3].

III. Results

A patient CTC dataset of size 512×512×413 was selected to test the above described ultra low-dose CTC pipeline for colon screening. A colon polyp of size 5 mm is centered at slice number 323 and indicated by an arrow in Fig. 1(a). All 413 slices of the volume CTC image were chosen to generate the line integrals or projection data. Each image slice generated a corresponding sinogram of fan-beam geometry.

After all the sinograms were generated from the 413 slices, a signal-dependent Gaussian noise was added according to the sinogram noise model of equation (1), simulating low-mAs acquisition protocols. The original CTC scan was acquired at a normal dose level of 100 mAs. The simulated ultra low-dose scan was at the level of 40 mAs. The simulated noisy scan was first reconstructed by the standard FBP algorithm (with Ramp filter at 100% Nyquist frequency cutoff). It can be observed that the polyp is blurred by excessive noise, see Fig. 1(b). The result of standard FBP reconstruction of the KL-PWLS restored sinogram is shown in Fig. 1(d). For comparison purpose, a conventional FBP reconstruction of the simulated noisy scan was also performed by carefully tuning the low-pass Hanning filter at an adequate cutoff frequency for a visually best result among the frequency range from 25% to 100% Nyquist frequency. The obtained result is shown by Fig. 1(c). It can be observed that the KL-PWLS sinogram restoration produces a better image quality than that of the Hanning filter, in terms of noise suppression and feature preservation. Their difference was further revealed by 3D endoscopic views using the V3D Colon-Module as follows.

All the reconstructed 3D images after Ramp filter (i.e., the standard FBP result), Hanning filter (i.e., the conventional FBP result) and KL-PWLS noise treatment respectively were further processed through the ECC pipeline and then fed into the V3D-Colon Module for both the construction of their corresponding virtual colon models and the navigation inside the constructed virtual models. Due to excessive noise presented in the images, the whole virtual colon model of the Ramp filter result could not be constructed by the V3D-Colon Module. The output was several separated colon segments. The whole virtual colon models from the Hanning filtered and the KL-PWLS treated results were successfully constructed and navigated by the V3D-Colon Module. The endoscopic views from which the polyp can be observed are shown in Fig. 2. It can be seen that detection of the polyp in the colon model from the KL-PWLS treated result is easier than that from the Hanning filtered result.

IV. Discussion and Conclusion

The KL transform provides a unique means to consider correlations among acquired data [4][6][9]. In this work, the KL transform was applied among neighboring slices of a 3D sinogram which was simulated from a patient CTC volume image by a slice-by-slice fashion in fan-beam geometry. For a routine clinical CT scan with helical acquisition mode, it is preferred to apply the KL transform directly on the 3D spiral-sampled sinogram data before interpolating the 3D data into 2D sinogram slices. The KL transform shall be directly applied to 3D helical CT sinogram of cone-beam geometry of flat panel detectors.

In summary, we have performed a simulation study to demonstrate a possible strategy for VC screening with ultra low-dose CT scans. Without noise suppression mechanism, the whole virtual colon model could not be constructed using the commercial V3D-Colon Module, which is dedicated to VC navigation mimicking the OC procedure. With an adequate noise reduction mechanism, the virtual colon model can be successfully constructed. The KL-PWLS noise reduction was shown to preserve more details on the polyp than the Hanning filter. This is confirmed by the endoscopic views on the polyp using the V3D-Colon Module navigation. This preliminary study indicates that an ultra low-dose CT-based VC is possible by the use of an adequate noise reduction strategy, such as the presented KL-PWLS sinogram restoration. It is expected that an ultra low-dose CT based VC could minimize the radiation risk and improve the compliance of colon screening recommendation and therefore reduce the morbidity of colon cancers. Further studies are needed which include (1) generating a large number (e.g., greater than 500) of ultra low-dose CT scans of the patient CTC data (with the 5 mm polyp) and performing a polyp detection task using receiver operating characteristic merit and (2) performing the polyp detection task on a large number of patient ultra low-dose CTC scans.

V. References

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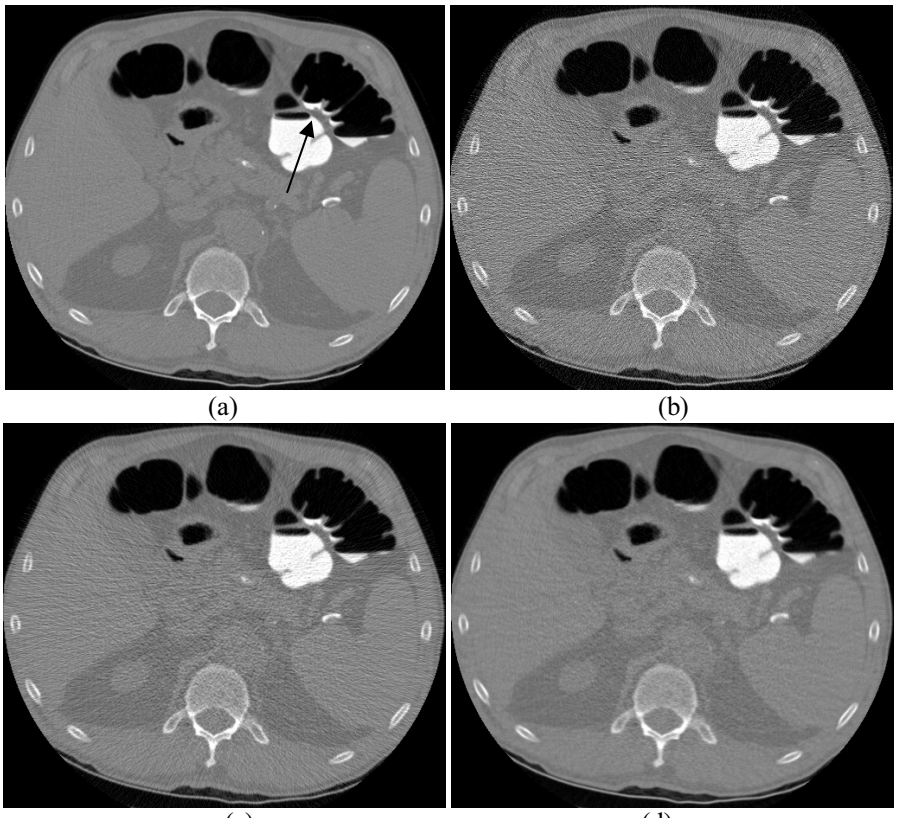
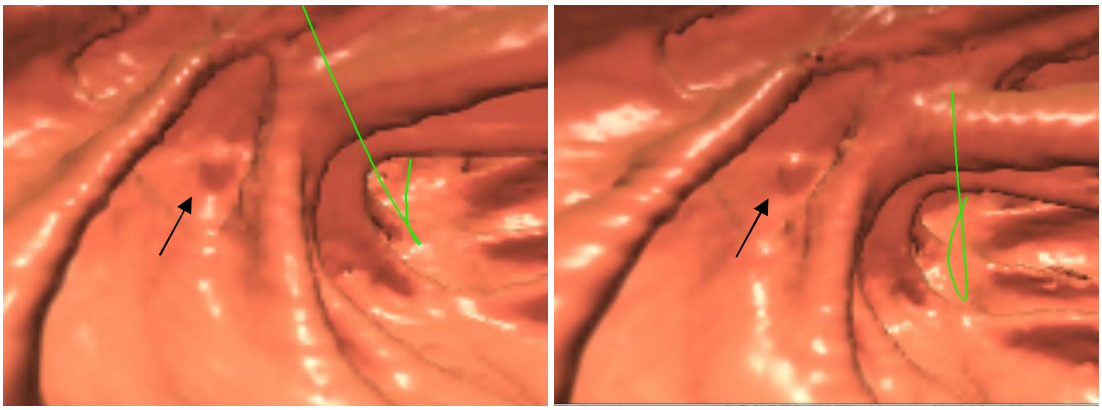


(a) (b) (c) (d)

Fig. 1: Illustration of one slice of a volume image: (a) from a normal dose scan (the arrow indicates the position of the polyp); (b) from standard FBP reconstruction of simulated ultra low-dose projection data; (c) from conventional FBP reconstruction of simulated ultra low-dose projection data, where the Hanning filter had a cutoff at 80% Nyquist frequency; and (d) from standard FBP reconstruction of simulated ultra low-dose projection data after KL-PWLS sinogram noise reduction was applied.



(a) (b)

Fig. 2: Endoscopic view of a polyp of 5 mm size: (a) from the result after noise reduction by the Hanning filter; (b) from the result after the KL-PWLS sinogram restoration. The arrows indicate the position of the polyp.