An Image Processing Technique for Automated Evaluation of the Relationship between Radiation Dose and Contrast-to-Noise Ratio in Dual Energy CT

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***Abstract*—This study presents an image processing technique for automated evaluation of the relationship between radiation dose and contrast-to-noise ratio in dual energy CT. Patient body size was calculated using region growing segmentation technique and used to estimate size-specific dose estimates. Homogeneous regions of interest on fat and contrast-enhanced liver area were localized, and contrast-to-noise ratio was automatically evaluated by our proposed method. Contrast-to-noise ratio turned out to be similar between 80 and 140 kVp for both pre- and post-contrast phase, whereas size-specific dose estimate was remarkably lower at 80kVp. Low tube potential can be recommended to reduce radiation dose while maintaining contrast-to-noise ratio.**

Keywords—CT, Image quality, contrast-to-noise ratio, size-specific dose estimates

I. Introduction

While computed tomography (CT) has been widely used with the advantages of 3-dimensional imaging of the human anatomy and rapid imaging capability, the main concern, radiation dose, limits more various applications. In order to optimize CT scan protocol, the relationship between image quality and radiation dose needs to be evaluated. However, image quality assessment has been usually carried out qualitatively in a subjective manner, which makes it difficult to apply to a large number of dataset with various scanners and protocols, and proper automated evaluation method was not established yet. High kVp settings were usually preferred with less noisy image, however, recent studies reported that CT scans with low kVp could produce images without contrast-to-noise ratio (CNR) degradation [1-3]. In this study, we developed computer assisted technique allowing an objective and efficient image quality and radiation dose assessment with expectation of accelerating protocol optimization by providing quantitative and reliable evidences.

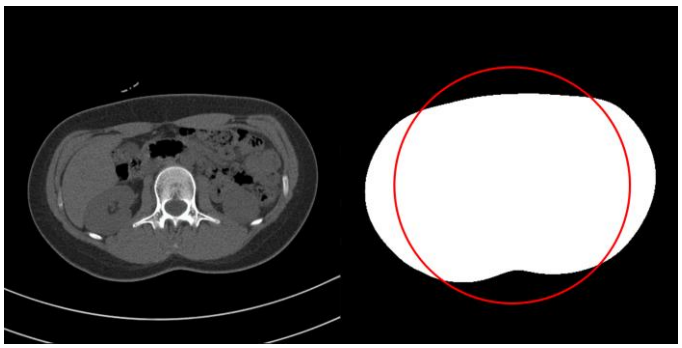

Fig. 1. Example axial CT image (*left*), and its segmented mask (*right*). Red circle shows an equal area with segmented body mask.

II. Materials And Methods

A. Materials

An abdominal CT data of single patient scanned with dual-energy protocol (80 and 140kVp) were obtained from picture archiving communication system in Seoul National University Hospital. Images were scanned from 128-row Multidetector CT (Somatom Definition, Siemens, Erlangen) with 3mm slice thickness. Automatic exposure control was applied showing 135-184 mAs in 80kVp scan, and 35-43 mAs in 140kVp scan, and a medium soft dual-energy kernel (D30f) was used for image reconstruction. Four contrast enhancement data of pre-contrast (PRE), early artery (EA), late artery (LA), and portal (PO) phases were used in this study.

B. CT Radiation dose metrics

Volume CT dose index, $CTDI_{vol}$, indicates the radiation dose of a poly-methyl methacrylate (PMMA) phantom rather than presenting the real amount of radiation dose exposed to patients with different body size. American Association of Physicists in Medicine (AAPM) task group (TG) 204 developed the conversion factor table which converts $CTDI_{vol}$ to size-specific dose estimates (SSDE) to give more accurate radiation dose considering the body size of each patient [5]. Three dimensional region growing technique was used to segment body region and remove CT table and patient cloths, and effective diameter (D_{eff}) was then estimated by drawing a circle which has an equivalent area with body region mask as depicted in Fig. 1. Derived from Monte Carlo simulation of 32cm PMMA phantom, conversion factor happened to be exponentially decreased as a function of

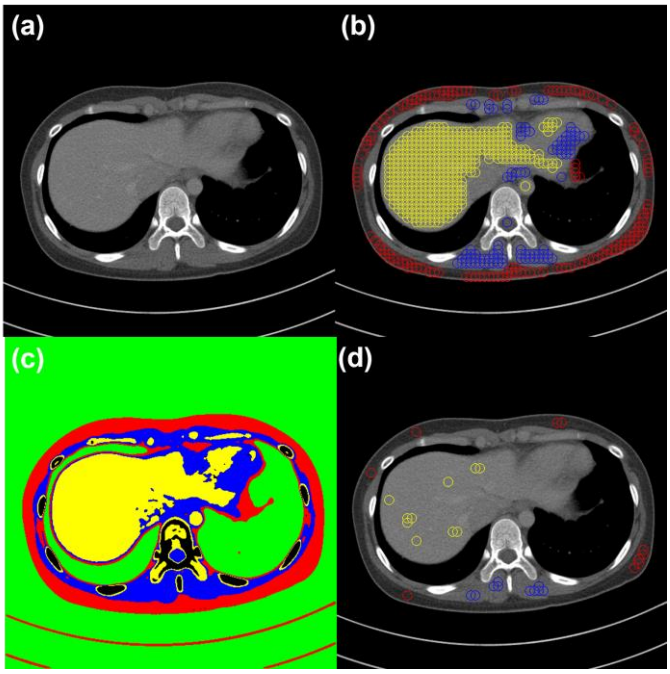

Fig. 2. (a) Example CT image, (b) initial candidate ROIs on predefined ranges, (c) tissue-specific initial mask, (d) selected top 10 homogeneous ROIs; fat with red circle, muscle with blue circle, and liver with yellow circle.

patient effective diameter on different X-ray tube voltage as in (1).

$$f(D_{eff}) = 4.378 \times e^{-0.043 \times D_{eff}} \quad (1)$$

SSDE was then yielded by multiplication of the $CTDI_{vol}$ with the conversion factor ($f(D_{eff})$) as in (2).

$$SSDE = CTDI_{vol} \times f(D_{eff}) \quad (2)$$

C. Homogeneous region localization

Each CT image was Gaussian filtered to reduce the effect of noise on homogeneity assessment. The initial candidate regions of interest (ROIs) with circular shape were placed with a criteria where the average intensities lie within predefined ranges (-200-0 HU for fat, 70-200 for contrast-enhanced liver region). With the guidance of AAPM TG 39 for raters to draw ROIs less than $1cm^2$ area, we set ROI size to fit on most subcutaneous fat regions by showing ROI area $0.65cm^2$, and same ROI size was applied in two different tissue regions. In order to exclude those pixels belonging to transition zones between different tissue structures, the sum of gradient magnitude was used to evaluate the homogeneity at each ROI. Among those candidate ROIs, top 10 homogeneous ROIs were selected for each tissue type, and example localization results were shown in Fig. 2.

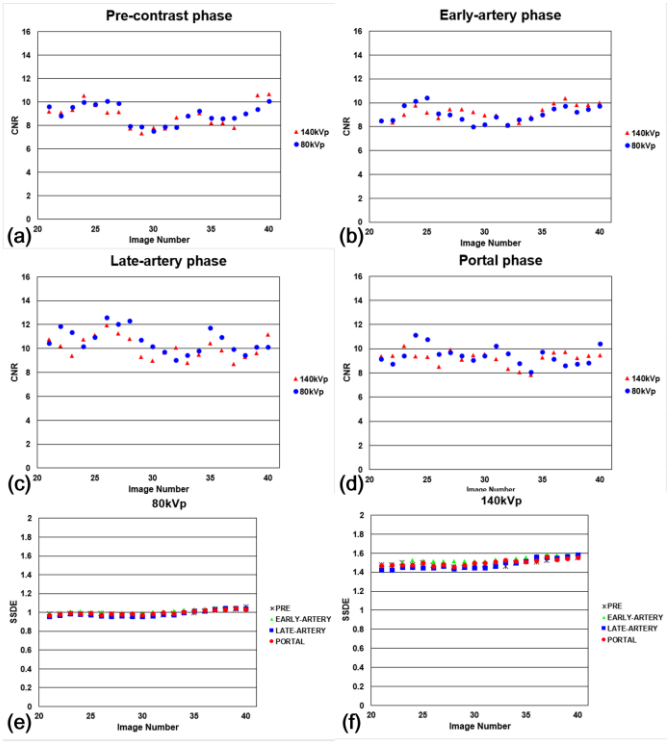

Fig. 3. CNR profiles at contrast-enhanced liver region with respect to fat for (a) pre-contrast phase, (b) early-artery phase, (c) late-artery phase, and (d) portal phase, and SSDE profiles at (e) 80 kVp, and (f) 140 kVp

D. Contrast-to-noise ratio

Representative mean intensity ($\bar{\mu}$) and standard deviation ($\bar{\sigma}$) were derived by averaging mean and standard deviation on localized homogeneous ROIs at each tissue region. CNR was defined as the absolute difference of $\bar{\mu}$ between fat and contrast-enhanced liver region over the standard deviation on fat as in (3). CNRs were compared on 4 contrast phases on 80 and 140kVp.

$$CNR = \frac{\left|\bar{\mu}_{CE\_liver} - \bar{\mu}_{fat}\right|}{\bar{\sigma}_{fat}} \quad (3)$$

III. RESULTS

SSDE and CNR on contrast-enhanced region with respect to fat of 80 and 140 kVp were compared for 4 contrast phases as shown in Fig. 3. Mean CNRs for 80 and 140 kVp were 8.96±0.83 and 8.90±0.99 for pre-contrast phase, 9.06±0.68 and 9.21±0.64 for early artery phase, 10.65±1.04 and 10.09±0.91 for late artery phase, and 9.43±0.77 and 9.22±0.60 for portal phase, respectively. CNR was shown to be much the same at 80 and 140 kVp for every contrast phase. Mean SSDE for 80 and 140 kVp were 0.99±0.03mGy and 1.49±0.03 mGy for pre-contrast phase, 1.01±0.02 and 1.53±0.03 mGy for early artery phase, 0.98±0.03 and 1.48±0.05 mGy for late artery, and 1.00±0.02 and

1.50±0.03mGy for portal phase, respectively. SSDE was shown to be significantly lower at 80 kVp than that at 140kVp. Low tube voltage protocol can be recommended by resulting in lower radiation dose without sacrificing CNR.

IV. CONCLUSIONS

Using our proposed technique including organ segmentation and homogeneous ROI localization steps, contrast-to-noise ratio and SSDE could be automatically evaluated. Due to the insufficient amount of photon strike to the detector in low kVp scan, images with low tube potential might appear noisier than with high tube voltage. However, recent studies have shown that imaging with low tube voltage provide comparable CNR while keeping patient exposure much lower [1-4]. Results of our study agree with those low kVp scan trends with expectation of about 30% dose reduction with 80 kVp protocol than with 140 kVp at comparable CNR level.

Our proposed techniques of homogeneous ROI localization needs to 1be modified to apply to more generalized imaging studies by incorporating more sophisticated organ segmentation technique such as atlas-based or probabilistic multi-organ segmentation. With test and validation procedures with larger data set, we expect our proposed technique has potential to be used as an automated tool in investigating the relationships between CNR and patients' dose, and thereby determining optimal scan parameters for various study applications.

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