Physica Medica 30 (2014) 385–390

Contents lists available at ScienceDirect

Physica Medica

journal homepage: http://www.physicamedica.com

Technical notes

Automatic exposure control at single- and dual-heartbeat CTCA on a 320-MDCT volume scanner: Effect of heart rate, exposure phase window setting, and reconstruction algorithm

Yoshinori Funama [a,*], Daisuke Utsunomiya [b], Katsuyuki Taguchi [c], Seitaro Oda [b], Toshiaki Shimonobo [d], Yasuyuki Yamashita [b]

[a] Department of Medical Physics, Faculty of Life Sciences, Kumamoto University, 4-24-1 Kuhonji, Kumamoto, Japan
[b] Department of Diagnostic Radiology, Graduate School of Medical Sciences, Kumamoto University, Kumamoto, Japan
[c] The Russell H. Morgan Department of Radiology and Radiological Science, The Johns Hopkins University School of Medicine, USA
[d] Department of Radiology, Kumamoto University Hospital, Kumamoto, Japan

A R T I C L E I N F O

Article history:
Received 13 August 2013
Received in revised form
16 October 2013
Accepted 22 October 2013
Available online 10 November 2013

Keywords:
Automatic exposure control (AEC)
Single- and dual-heartbeat CTCA
Image noise
Radiation reduction
320-Detector volume scanner

A B S T R A C T

Purpose: To investigate whether electrocardiogram (ECG)-gated single- and dual-heartbeat computed tomography coronary angiography (CTCA) with automatic exposure control (AEC) yields images with uniform image noise at reduced radiation doses.
Materials and methods: Using an anthropomorphic chest CT phantom we performed prospectively ECG-gated single- and dual-heartbeat CTCA on a second-generation 320-multidetector CT volume scanner. The exposure phase window was set at 75%, 70–80%, 40–80%, and 0–100% and the heart rate at 60 or 80 or corr80 bpm; images were reconstructed with filtered back projection (FBP) or iterative reconstruction (IR, adaptive iterative dose reduction 3D). We applied AEC and set the image noise level to 20 or 25 HU. For each technique we determined the image noise and the radiation dose to the phantom center.
Results: With half-scan reconstruction at 60 bpm, a 70–80% phase window- and a 20-HU standard deviation (SD) setting, the imagenoise level and -variation along the *z* axis manifested similar curves with FBP and IR. With half-scan reconstruction, the radiation dose to the phantom center with 70–80% phase window was 18.89 and 12.34 mGy for FBP and 4.61 and 3.10 mGy for IR at an SD setting SD of 20 and 25 HU, respectively. At 80 bpm with two-segment reconstruction the dose was approximately twice that of 60 bpm at both SD settings. However, increasing radiation dose at corr80 bpm was suppressed to 1.39 times compared to 60 bpm.
Conclusion: AEC at ECG-gated single- and dual-heartbeat CTCA controls the image noise at different radiation dose.

Introduction

Multidetector computed tomography coronary angiography (CTCA) with electrocardiogram (ECG) gating is an accurate non-invasive method to evaluate coronary artery disease [1–4]. Potential applications of CTCA raise concerns regarding radiation exposure. The radiation dose is reduced by techniques that use step-and-shoot prospective ECG-gating, retrospective ECG-gating with tube current modulation (TCM), tailoring of the tube current, low-tube voltage, iterative reconstruction, and ensuring the smallest scan range [5–9]. Wide coverage CTCA with 320-detector row volume acquisition also promises to reduce the radiation dose [10–14]; in clinical CTCA studies it has been reported to deliver an effective dose of less than 1 mSv [15].

Automatic exposure control (AEC) for CTCA scanning has been introduced on 320-multidetector (MD) CT volume scanners. The volume scanner allows axial volumetric scanning up to 16 cm in the *z*-axis in a single gantry rotation with no table movement. Different from retrospectively ECG-gated methods with TCM, the tube current is automatically adjusted to reflect the attenuation in patients with different body shapes and densities during one gantry rotation. In combination, wide coverage CTCA and AEC can be expected to further decrease the radiation dose while maintaining the image quality. However, the effect of heart rate variations, of the exposure phase window in the R–R interval, and of the reconstruction

* Corresponding author. Department of Medical Physics, Faculty of Life Sciences, Kumamoto University, 4-24-1 Kuhonji, Kumamoto 862-0976, Japan. Tel./fax: +81 96 373 5455.
E-mail address: funama@kumamoto-u.ac.jp (Y. Funama).

1120-1797/$ – see front matter © 2013 Associazione Italiana di Fisica Medica. Published by Elsevier Ltd. All rights reserved.
http://dx.doi.org/10.1016/j.ejmp.2013.10.003

technique on the relationship between the image quality and radiation dose at ECG-gated single- and dual-heartbeat CTCA with AEC remains unknown.

We investigated whether ECG-gated single- and dual heartbeat CTCA with AEC on a 320-MDCT volume scanner yields images with uniform image noise at reduced radiation doses. Our studies involved combinations of different heart rates, different reconstruction techniques (filtered back projection [FBP] or iterative reconstruction [IR]), and different exposure phase window settings.

Materials and methods

As no patients or animals were scanned and no clinical images were used, this study did not require institutional review board approval.

Phantom

We used a commercially available chest CT screening phantom with a dose meter hole in its center (LSCT001, Kyoto Kagaku Co., Tokyo, Japan). The external dimensions of the phantom were chest circumference 930 mm, height 370 mm. The chest phantom was based on projection data obtained from CT chest scans of a patient 175 cm in height and 75 kg in weight. A radiation dosimeter was inserted along the *z* axis of the 13-mm diameter dose meter hole.

ECG-gated single- and dual-heartbeat CTCA

Using the anthropomorphic chest CT phantom we performed prospectively ECG-gated single- and dual-heartbeat CTCA on a second-generation 320-MDCT volume scanner (Aquilion ONE ViSION; Toshiba Medical Systems, Otawara, Japan). The scan parameters were detector configuration, 320 $\times$ 0.5 mm (detector collimation); slice thickness, 0.5 mm; gantry rotation time, 0.275 s; display field-of-view (FOV), 240 $\times$ 240 mm; matrix, 512 $\times$ 512. An ECG was acquired during volume scanning at heart rates of 60 and 80 bpm using the demo mode for cardiac CT. Due to temporal resolution issues our CT scanner changed from half-scan reconstruction, which uses projection data over less than one rotation for a temporal resolution of 0.138 s, at 60 bpm to two-segment reconstruction at 80 bpm. The tube voltage was 120 kVp. For single- and dual-heartbeat CTCA we used a new AEC technique which can be explained as follows: First, the X-ray attenuation within the scan range was measured on a patient's scout view image. Second, the water-equivalent thickness of the object was calculated from x-ray attenuations. Finally, the tube current value was obtained using the relationship between setting image noise level and maximum value of image noise of water-equivalent object given the thicknesses. Therefore, decided tube current value was the same value along *z*-axis of the patient for scanning whole heart with no table movement. The image noise level (standard deviation [SD] setting) was set at 20 and 25 Hounsfield units (HU) on the CT scanner. The minimum and maximum tube current thresholds for AEC were set at 10 and 750 mA, respectively, before CTCA scanning. The AEC for cardiac volume scanning also corresponded with iterative reconstruction (IR, adaptive iterative dose reduction 3D [AIDR3D], Toshiba Medical Systems, Otawara, Japan) when the standard level of blending was used to reduce the radiation dose [16,17]. For cardiac volume scanning we set the exposure phase window (padding window) to 75%, 70–80%, 40–80%, and 0–100%. Two different synthesized heart rates, 60 and 80 bpm, were used, which prospectively set ECG-gated image reconstruction algorithm to use: halfscan algorithm for 60 bpm and 2-segment reconstruction for 80 bpm. The default AEC technique does not take into account the heart rate, thus, the radiation dose may increase with 2-segment reconstruction. The AEC technique has an option to correct for overlapped projections of two heart beats in the 2-segment method. Thus, six scans were performed in total, at two image noise level, 20 and 25 HU, and with three algorithms, halfscan algorithm for 60 bpm, 2-segment without the overlap correction for 80 bpm ("80 bpm"), and 2-segment with the correction for 80 bpm ("corr80 bpm"). Images were synchronously reconstructed with ECG data using FBP and IR algorithms and a coronary standard kernel/filter (FC11). The center of the imaging window width was set at 75% of the R–R interval.

Image noise measurements

One author (Y.F.) measured the image noise for each scanning technique at the mediastinal portion of the chest CT phantom. The noise index was the SD of the pixel values within a 20-mm-diameter circular region of interest (ROI). We acquired 185 consecutive images in the *z* direction at each scan, performed 3 scans, and then calculated the mean image noise value.

Measurement of the radiation dose delivered to the center of the phantom

To measure the radiation dose delivered to the center of the phantom we used a 10-cm pencil-shaped ionization chamber (model 20 $\times$ 5 10.3 CT; Radcal Corp., Monrovia, CA) connected to a Radcal (model 9015) dosimeter. The phantom harbored a 13.0-mm diameter cavity in its center. The ionization chamber was inserted into the cavity and the dose delivered to the center by the identical AEC scanning technique was measured. The measurements were repeated 3 times and the mean dose was calculated for each of the 24 scan settings (60 bpm: 2 reconstruction methods $\times$ 4 exposure phase windows $\times$ 3 scans) and the 48 scan settings (80 bpm: 2 reconstruction methods $\times$ 4 exposure phase windows $\times$ 6 scans) with ($n = 3$) or without ($n = 3$) correction by 2-segment reconstruction.

Statistical analysis

To assess the image noise for FBP and IR we used the independent samples *t*-test (Student's *t*-test). If there was no significant difference in both image noises, post hoc power ($1-\beta$) analysis between the groups using standardized mean differences and sample size and a level of 0.05 (α) was also applied with G*Power software (version3.1.; University of Düsseldorf, Düsseldorf, Germany). We also used Bland–Altman [18] analysis to determine the image noise with 2-segment reconstruction at 80 bpm and corr80 bpm. Differences of $p < 0.05$ were considered statistically significant. Calculation of the student's *t*-test and Bland–Altman analysis was with a statistical software program (MedCalc, Mariakerke, Belgium).

Results

Image noise

At 60 bpm, with half-scan reconstruction, a 70–80% phase window, and the 20-HU SD setting, the image noise level and variation along the *z* axis (from top to bottom) manifested similar curves with FBP and with IR (Fig. 1a). The image noise for FBP and IR increased from the upper cardiac level to the lower cardiac level (range 12.20–21.17 HU [FBP] and 13.47–19.27 HU [IR]). In addition, mean image noise and 95% confidence interval (CI) for FBP and IR were 14.92 HU and 15.33 HU and 14.52–15.32 HU and 14.99–15.67 HU. There was no statistically significant difference with respect to