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Preliminary Experimental Results on Controlled Cardiac Computed Tomography: A Phantom Study

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Abstract

In this paper, we present the preliminary experimental results on controlled cardiac computed tomography (CT), which aims to reduce the motion artifacts by means of controlling the x-ray source rotation speed. An innovative cardiac phantom enables us to perform this experiment without modifying the scanner. It is the first experiment on the cardiac CT with speed controlled x-ray source. Experimental results demonstrate that the proposed method successfully separates the phantom images at different phases (improve the temporal resolution) though controlling the x-ray speed.

Keywords

cardiac computed tomography; controlled gantry rotation speed; image reconstruction

I. Introduction

Cardiovascular disease ranks first in the leading cause of death in United States for many years [1]. In 2005, there were 652,000 Americans died of cardiovascular diseases, in other words, in every three death, one was caused by cardiovascular disease. It is also estimated that one in five Americans has some form of cardiovascular diseases (<http://www.americanheart.org/>), and the estimated cost of cardiovascular diseases has reached \$368.4 billion in 2004. To that end, early detection and prevention of cardiovascular diseases is vitally important.

To diagnose the cardiovascular disease, an accurate and clear cardiac image is essential. However, imaging the heart is challenging due to its voluntarily rapid motion, and both temporal and spatial resolution are crucial to the image results. High spatial resolution means clear detailed images; while high temporal resolution is greatly expected to reduce the motion artifact. Currently, CT (Computed Tomography), Electron-beam Computed Tomography (EBCT) and MRI (Magnetic Resonance Imaging) are the three main imaging modalities used for cardiac imaging.

Electron-beam CT (EBCT) is one mode of medical x-ray CT, which is specially designed for dynamical images [2,3,4]. EBCT has a non-mechanical X-ray source, which allows for

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image acquisition in 50~100ms. However, 1) the EBCT does not produce the high quality images as conventional CT does because its x-ray spot is not sufficiently intensive and 2) the EBCT scanner is much too expensive, which limits its accessibility and affordability.

MRI is another popular imaging tool. It is non-invasive and no radiation is involved. However, its application is greatly limited on cardiac imaging by its intrinsic nature that temporal resolution is inversely related to spatial resolution, which means that spatial resolution and temporal resolution cannot be improved simultaneously on a MRI machine. Besides, the resolution of MRI is not high as that of the CT images [5], and MRI is not suitable for patient with implants or has claustrophobia.

With the advent of Multi-row Detector CT (MDCT), the spatial and temporal resolution of MDCT has been greatly improved. Meanwhile, researchers have been trying many technologies to improve the temporal resolution for cardiac image: 1) Half-scan reconstruction: half-scan (HS) reconstruction [6] algorithm uses scan data from $>180^\circ$ gantry rotation (180~250ms) to generate one single axial image, therefore, it improves the temporal resolution by less than two factors. 2) Dual-source CT: it uses two sources and two detectors at the same time. Basically, this technology increases the temporal resolution by at least two times and provides the full details of the heart [7]. 3) ECG gating/multi-segment reconstruction: The idea of ECG gating/ multi-segment reconstruction is to record the ECG signal simultaneously while scanning the heart [8,9], thus, the projection data with less heart motion can be identified through the ECG signal. By rearranging those projection data (for one phase) to form a 360° (or $>180^\circ$) data from several consecutive/non-consecutive cardiac cycles, technicians can reconstruct an image with less motion artifacts.

All of above three technologies does not guarantee the elimination of the heart motion artifacts. The first and the second methods try to shorten the acquisition time, which is still greater than 150 ms. As for the ECG gating/multi-segment reconstruction, there are however two major problems [10,11]. First, there is no guarantee that a complete projection data (360°) for a specific phase will be obtained during the scan. Prolonging the scan time does not necessarily complete that data set either; besides, it increases the radiation dose [12,13]. Second, even in the favorable cases, retrospectively reconstructed cardiac images still suffer from substantial motion blurring because in practice each projection sector covers a projection angular range of a substantial length. Within such an angular range, the heart will move appreciably, especially when it is not in a relative stationary phase.

Now we come to the question: Is there a solution that can eliminate motion-induced image artifacts for MDCT without any major hardware change? The answer is positive. In this paper, we propose a new approach, which is completely different from any existing technique. This approach integrates modern control and cardiac imaging techniques, that is, by adaptively controlling the imaging device (x-ray source) according to the real-time analysis and prediction of an individualized respiratory and cardiac motion functions, little artifacts would be left.

Integrating control concept into imaging field is not new, and we have reported many results on that. Controlling x-ray source to reduce the heart motion artifact for cardiac CT was shown in [14,15], whose concepts will be demonstrated in this paper. Bolus chasing CT angiography is another example of control application in imaging field, which has been conceptually and experimentally demonstrated in [16,17,18].

The organization of the paper is as follows: General schemes are illustrated in section 2, where we will restate the problem and control schemes for controlled cardiac CT. In section 3, we will describe our experimental setup in detail, including the phantom design and scan mode on our in-house Siemens CT scanner. Experimental results will be shown in section 4

together with the comparison for no control cases. Finally, discussion and conclusion are given in section 5.

II. General schemes

Problem statement

Computed tomography is a powerful non-invasive evaluation technique for producing two-dimensional and three-dimensional cross-sectional images of an object. The idea of the CT is illustrated in Fig. 1: the object/patient is in the gantry, and x-ray source and the detector rotate around the object/patient. The detector collects the projection data for every angle, and the computer uses those projection data to reconstruct the cross-sectional images. The spatial resolution of the image depends on the number of projection data and detector settings. During the CT scan, the object/patient has to keep still; otherwise, motion artifacts, which cause unsatisfactory images, will be produced.

The heart scan is more challenging because 1) the heart contracts and expands rapidly all the time, 2) the difference of the heart shape/volume is substantial during its motion. One possible solution to increase the temporal resolution is to increase the x-ray source and detector rotation speed so that all the projection data are obtained with little change of the heart motion. The current state of art gantry rotation speed, 0.33 seconds per rotation, has almost reached its physical limit. However, with this rotation speed (temporal), it is not enough to image the heart structure clearly.

Another way to increase the temporal resolution is to scan the object (heart) for several cycles and select and arrange the angle projection data properly to reconstruct separate images for different phases. This requires accurate computation and deliberate control of gantry rotation, which ensures that projection data for a specific phase is complete and the scan time is minimized.

To state this problem in a precise way, we define a few terminologies.

- i. Let $v(t)$ represent the geometry status (phase) of the object at time t . For example, if we want to scan the patient heart, $v(t)$ will represent the heart volume at time t . The phase $v(t)$ has to be finite, i.e. $v(t) \in \{v_1, v_2, \dots, v_p\}$.
- ii. Let $a_{i,l} = \frac{(l-1)}{p} 2\pi, l=1, 2, \dots, L_i$ be L_i evenly spaced angles to collect the projection data for phase v_i .
- iii. Let $s(t)$ represent the x-ray source angle at time t , and $s(0) = 0$. It is a function of time and its rotational velocity is assumed to be controllable.

The controlled cardiac CT problem is stated as follows: *Find a source angle profile $s(t)$ as a function of time t so that for every phase v_i at every angle $a_{i,l}$, there exists a corresponding time t such that $s(t) = a_{i,l}$. In such a way, through rearranging the projection data, the object's CT images at different phases can be reconstructed with no motion artifacts.*

Control scheme

The control schemes for controlled cardiac CT are given in paper [14]. In this paper, we will focus on the experimental results; therefore, we just restate the control scheme results for periodic heart motion. As for non-periodic cases, it will be discussed later.

Control scheme for a periodic heart motion

Assume the heart volume $v(t)$ is periodic with period P , that is,

$$v(t)=v(t+P) \forall t$$

and assume the gantry rotation time for one circle is P_c .

Theorem [14]: Given $P, t_i, \theta_i + \frac{k_{i,l}\bar{p}+w-1}{p}2\pi$ and $p = \bar{p}\bar{m}$ for some $\bar{p} < p$. $s(t) = \frac{2\pi}{P_c}t$ covers all the projection angles $\theta_i + \frac{k_{i,l}\bar{p}+w-1}{p}2\pi$ at times $t_i + (l-1)P + (w-1)\frac{P_c}{p}$ if and only if

$$\frac{P}{P_c} = m + k\frac{\bar{p}}{p} \quad (1)$$

where $w \in [1, 2, \dots, \bar{p}]$, and $m \geq 0$ is any integer and $k \in [1, 2, \dots, \bar{m}-1]$ and is not a multiple of any non-trivial divisor of $\bar{m}$.

The above theorem does not give a specific solution for controlled cardiac CT; however, it provides all the solutions under the situation for known periodic heart motion. What the controller needs to do is that first judge if condition Eq. (1) is satisfied. If it is satisfied, there is no need to change the gantry speed and just collect the projection angle data and reconstruct the image; otherwise, the gantry rotation speed will be varied accordingly so that Eq. (1) is met.

III. Experimental design

Cardiac phantom

An innovative cardiac phantom is designed and built for this cardiac CT experiment. It is made of plastic with low density, and its dimensions are given in Table 1. We transform the two-/three- dimensional motion of the heart into a one-dimensional motion, in such a way; the “heart volume” can be easily controlled. The idea is as follows: the phantom is a cone-like (Fig. 2) object with its cross-sections along its height being different size of ellipses. The phantom will be placed horizontally (Fig. 3 right), with its axis being at the center of the gantry. Moving the phantom back and forth changes the phantom’s cross-sectional area under the x-ray, which mimics the heart expansion and contraction. The point is that in this design, the heart volume (area of ellipse) is directly related to the position of the phantom under the x-ray.

The phantom will be tested on an in-house Siemens SOMATOM Volume Zoom CT scanner (Fig. 3 Left), which has four-detector rows. Due to the proprietary problem, we are not allowed to change the gantry speed during the scanning, and the gantry speed has to be 0.5 sec, 1 sec, or 1.5 sec per rotation. To that end, the only choice to control the x-ray source rotation is to rotate the phantom itself and change the relative rotation speed between the x-ray source and the phantom.

Motion system

Now we need to drive the phantom linearly and rotationally, and both motions are realized through two stepping motors, ASM46AA, from orientalmotor.com. The motors have built-in driver ASD13A-AP, and can be directly controlled by PC through a communication cable. The resolution of the motor is 1000 pulse/rotation or 0.36° .

Remark

The gantry center and the rotation axis of the phantom have to be concentric because we are rotating the phantom instead of the x-ray gantry, and all the projection angle data sets have to be transformed back for reconstruction. Due to the manufacturing, the bench of the experiment vibrates while the phantom is moving too fast. To that end, we have to limit its velocity.

Scan mode

Sequential scan is used for this experiment for better images. First, sequential scan collects data at one specific position, and secondly, spiral scan needs the z-direction movement, which makes it complicated since we will translate the phantom inwards and outwards of the gantry to mimic the heart motion. Ideally, we want a continuous scan and select the projection angle data from the saved raw data. But there is no such a mode in the given CT. We choose to use Head Seq. for scan mode despite it has a shortcoming: 1.5 seconds idle time between two scans. To that end, we will not apply the aforementioned theory directly to this experiment. Instead, we just try to demonstrate the concept of controlled cardiac CT. Table 2 shows the scan parameter settings.

Remark

CareVision scan mode can continuously scan the object with/without moving the CT table; however, this model does not provide the raw data. To that end, we choose the head Seq. mode. By setting the feed to be zero, we can scan one specific cross-section continuously.

No. of Scans decides the total scan time, radiation exposure time, which equals to scan time multiplied by No. of scan. In this preliminary experiment, we cannot turn on and off the x-ray at our will; therefore, the radiation exposure is limited by reduction of the No. of scan.

To better demonstrate the performance of the controlled cardiac CT, we use full scan reconstruction algorithm. Half-scan algorithm is also applicable in this case, and we will discuss it later. The reconstruction procedure is listed as follows.

1. Select projection angle data from different scans to form 360° data sets for different phases (two in the experiment).
2. Transform the projection angle data. Since the gantry speed is varied, and the projection data is not for a specific gantry rotation speed (0.5 sec, 1 sec or 1.5 sec per rotation), it is required to transform the projection data before the reconstruction.

Suppose the gantry has a constant rotation speed ω_1 and the phantom has a constant rotation speed ω_2 ($\omega_1 > \omega_2$ with the same direction). Then,

$$\begin{aligned} D(s_c) &= D(\omega_1 t) \\ D(s_r) &= D((\omega_1 - \omega_2)t), \end{aligned}$$

where $D(s_c)$ and $D(s_r)$ represent the compressed projection at angle s_c and the regular projection at angle s_r , respectively. For a full scan algorithm, s_r must cover the range of 2π , thus the minimum scanning time is $t_{\min} = 2\pi/|\omega_1 - \omega_2|$. Then, the compressed projection should be extended to a range of 2π for the scanning time $t_{\min}$. That is

$$D(s) = D\left(\frac{\omega_1}{\omega_1 - \omega_2}t\right)$$

3. Reconstruct the CT image for each phase using the program provided by Siemens.

IV. Experimental results

Phantom linear motion

The phantom linear motion trajectory is shown in the Fig. 4. It has two static phases, *phase 1* at 0 cm (somewhere middle of the phantom) and *phase 2* at 4.5 cm, and one transition phase. In this experiment, we will only reconstruct phase 1 and phase 2. As for transition phase, there are infinite images at different position between 0 cm and 4.5 cm, which will not be discussed in this experiment. For simplicity, we set the phantom linear motion to be periodic, however, the static time for phase 1 and 2 are different (1 sec and 0.75 sec, respectively). The reason to set this linear motion trajectory is two-fold: 1) we want to show that when period of the phantom motion is integer times of the gantry rotation speed, it is impossible to reconstruct a clear image without controlling the gantry speed; and 2) 1.5 sec is lowest gantry rotation speed, combined with intermittent scan mode, 4.5 sec period is the best we can do to demonstrate the concept.

No control case

The gantry rotation time is set as 1.5 sec /rev, so it is impossible to obtain the complete projection data for each phase under one scan. Hence, without changing the rotation speed, the CT images of the phantom are mixed with *phase 1*, *phase 2*, and transition phases. Fig. 5 shows the phantom images, which have lots of motion artifacts on the boundaries. The original images are 512×512 , and for better presentation, we show the images with 256×256 .

The phantom motion period is 4.5 seconds, which is three times of the gantry rotation time. According to the theorem, it is impossible to rearrange the projection data to obtain the 360° data for reconstruction purpose. Fig. 6 shows that with 1.5 sec/rev, the useful projection data for *phase 1* only covers $0 \sim 240^\circ$, and $180 \sim 360^\circ$ for *phase 2*.

Controlled Cardiac CT

Now we are at the situation to obtain clear CT images for the cardiac phantom at phase 1 and phase 2. Assume the CT gantry rotation cannot reach 1 sec/rev, which means temporal resolution cannot be achieved through increasing CT gantry speed. To that end, the only possible solution is to increase the No. of scan for the phantom, and select and rearrange the appropriate projection data for reconstruction.

Unfortunately, as we mentioned before, the current existing rotation speed cannot satisfy the reconstruction requirement. To that end, we have to change/control the gantry rotation speed, which is realized through rotating the phantom about the axis centered at the gantry center.

To illustrate this method, we use two rotation speed, -4 sec/rev and 12 sec/rev. The “minus” sign means that the phantom rotates in the opposite direction of the gantry. We want to show that controlled cardiac CT works well even the gantry rotation speed is reduced, which benefits the CT with poor temporal resolution. Fig. 7 and Fig. 8 show the selection of angle data for different phantom rotation speed, -4 sec/rev and 12 sec/rev, respectively. It is clear that, through varying the phantom rotation speed (gantry speed), there are a lot of choices to obtain the complete projection data for phase 1 and 2. Fig. 7 and Fig. 8 are just for demonstration. For instance, in Fig. 7, complete projection data for phase 1 can be obtained through scan 1 and 4, while scan 3 and 9 completed the 360 degree projection data for phase 2.

Fig. 9 and Fig. 10 show the reconstructed CT images of phantom for phase 1 and phase 2 with phantom speed -4 sec/rev and 12 sec/rev, respectively. Obviously, both two experiments successfully separate the phase 1 and phase 2, though there are some blurs at the boundary of the images, which may caused by 1) non-concentricity of the rotation axis and gantry center and 2) experiment vibration. As for the concept demonstration, this is more than enough.

Image quality is influenced by many factors. One major factor is the non-concentricity of the rotation axis and gantry center. It is not a problem when the gantry is still. However, while the phantom is rotating, the non-concentricity of the rotation axis and gantry center makes the central line on the detector change (see Fig. 11), and then the projection data cannot be directly used for reconstruction. We find it is very difficult to make the rotation axis exactly overlap with the gantry's center and the reconstructed images are very sensitive to the non-concentricity, so a center correction method is used to compensate the deviation of the central line. We estimate the center of the phantom at different phases, calculated the deviation of the central line, and then transform the projection data according to the new central line. Theoretically speaking, the center correction method will eliminate all the artifacts caused by the non-concentricity, however, due to the vibration of the gantry, the center of the phantom will slightly change each time, thus the artifacts always exist.

VI. Discussions and conclusions

The reconstructed images still have some motion artifacts, which are induced by the phantom rotation and the vibration of the experiment setup. Rotating the phantom instead of controlling the gantry speed strictly requires that the rotation axis and the gantry center be concentric. Otherwise, the projection data will be different from that only rotating the x-ray source. However, superposing the rotating axis and gantry center is extremely hard because the gantry center is in the air and there is no base point for reference. Vibration of experiment setup causes the phantom to shake, especially when it stops and starts. On the other hand, the motion artifacts is greatly reduced compared to the passive case even with the presence of concentric and vibration problem.

In this paper, we only show a very simple case, periodic scenario, for controlled cardiac CT. The control scheme is very simple, rotating the phantom in the opposite/same direction of the gantry rotation. However, the results are substantial: the reconstructed images almost eliminate the linear motion artifacts.

In the real cases, the cardiac CT is complicated with non-periodicity, which needs to be estimated on line. We would like to show the experimental results for that, but at this stage, we just want to demonstrate that the proposed concept works well.

Image reconstruction method, which is closely connected with the control scheme, can be improved also. In this experiment, full scan algorithm was used for the purpose of better noise tolerance. In fact, we can choose many other reconstruction algorithms for different purposes. For example, if the projection data we collected cannot cover the range of 2π , half scan algorithm or super-short-scan algorithm [19] should be considered. Both of these two algorithms can reconstruct the image by using projection data less than 2π , but the latter is more flexible in the distribution of the projection data, i.e. the projection data can be discontinuous. For many other cases, we are only interested in peripheral or central area of the object, and then BPF algorithm [20] can be used. Even in the case that projections are truncated on one side or two sides, we can use interior tomography [21,22] to reconstruct the image. All of these algorithms can be applied to our controlled cardiac CT, according to the design of the control system.

In this paper, we present the experimental results of the controlled cardiac CT, which aims to reduce the motion artifacts through controlling the gantry rotation speed. Innovative cardiac phantom is tested on our in-house CT scanner. Experiment results demonstrate that this method successfully separate the phantom images at different phases (improve the temporal resolution) without necessarily increasing the gantry speed. It is the first experiment on the cardiac CT with gantry speed controlled, though it is realized through rotating the phantom. Future work will focus on the non-periodic heart motion.

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Fig. 1.
Illustration of CT

Fig. 2.
Conceptual Cardiac phantom

Fig. 3.
Left) Siemens SOMATOM Volume Zoom CT scanner, and Right) experimental setup.

Fig. 4.
Linear motion of the phantom

Fig. 5.
CT images of the cardiac phantom with only linear motion. Those two images are for different starting time, but both show that they are mixed with the different phases.

Fig. 6. Projection data for phase 1 (top) and phase 2 (bottom) for total 10 scans for phantom with linear motion in Fig. 4 and no rotation motion. Red box represents the time when the x-ray is on, and the blue line represents the angle data that can be used for reconstruction for corresponding phases.

Fig. 7. Projection data for phase 1 (top) and phase 2 (bottom) for total 10 scans for phantom with linear motion in Fig. 4 and rotation speed -4sec/rev . Red box represents the time when the x-ray is on, and the blue line represents the angle data that can be used for reconstruction for corresponding phases.

Fig. 8. Projection data for phase 1 (top) and phase 2 (bottom) for total 12 scans for phantom with linear motion in Fig. 4 and rotation speed 12 sec/rev. Red box represents the time when the x-ray is on, and the blue line represents the angle data that can be used for reconstruction for corresponding phases.

Fig. 9.
Reconstructed CT images for cardiac phantom of phase 1 (Left) and phase 2 (Right).
Phantom rotation speed is -4 sec/rev.

Fig. 10.
Reconstructed CT images for cardiac phantom of phase 1 (Left) and phase 2 (Right).
Phantom rotation speed is 12 sec/rev.

Fig. 11. Illustration of Non-concentricity of the rotation axis and gantry center. Point A and B are the centers of the gantry and the phantom, respectively. $\Delta\theta$ indicates the deviation of the central line.

Table 1

Dimensions of the cardiac phantom

Top surface long axis	2cm	Top surface short axis	1cm
bottom surface long axis	10cm	bottom surface short axis	5cm
Height	15cm	Weight	≈ 250 g

Table 2

CT scan parameter settings

Scan mode	Head Seq.	Slice	$1 = 2 \times 0.5 \text{ mm}$
Scan time	1.5 sec	Cycle time	3.0 sec
No. of Scans	5	Feed	0 mm