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Original paper

A novel design of ultrafast micro-CT system based on carbon nanotube: A feasibility study in phantom ☆

Zhicheng Zhang [a,b], Shaode Yu [a,b], Xiaokun Liang [a,c], Yanchun Zhu [a,*], Yaoqin Xie [a,*]

[a] Shenzhen Institutes of Advanced Technology, Chinese Academy of Sciences, Shenzhen, China
[b] Shenzhen College of Advanced Technology, University of Chinese Academy of Sciences, Shenzhen, China
[c] School of Information Engineering, Guangdong Medical College, Dongguan, China

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A B S T R A C T

Artifacts induced by respiratory motion during routine diagnosis severely degrades the image quality. The increase of scanning speed plays an important role to avoid motion artifacts. Limited to the mechanical structure of conventional CT, the increase of gantry rotational speed is unsustainable and a more feasible way is to increase the number of X-ray sources and detectors like the dual-source CT. This paper focuses on high-speed scanning CT and proposes a novel ultrafast micro-CT (UMCT) system based on carbon nanotube (CNT). At each exposure position, all of the X-ray sources are fast activated by turns and the flat-panel detectors collect the corresponding projection data. Then, the gantry will be contrarotated 40° to prepare for the next exposure until the rotation covers full 360°. Because each exposure is very fast, the organ motions of *in vivo* human body can be greatly reduced. This paper introduces the UMCT system design, image reconstruction algorithm and experimental results. Simulation experiment was also carried out on UMCT system. The result validated the feasibility of the UMCT system.

1. Introduction

3D microscopy which has a high spatial resolution can be revealed non-destructively for fine-scale internal structures is X-ray-based three-dimensional (3D) imaging for a limited field of view (FOV) [1,2]. It has been widely used in small animal imaging and plays an important role in the research of human organ, exploration of disease mechanisms and effectiveness of drug assessment [3]. As to currently commercial scanners, it is still a big challenge to perform *in vivo* micro-CT on small animals because of their much faster physiological motion than that of human. Motion artifacts severely degrade the micro-CT imaging quality and significantly affect the effectiveness of disease diagnosis [4].

Conventional micro-CT uses a hot cathode based X-ray source. According to the theory of thermionic electron emission, the electron source must be heated above 2000 °C in order to allow free electrons to escape from its surface [5], thus it is impossible for hot cathode based X-ray source to be switched on and off instantaneously. Using field-emission sources to replace thermionic sources was suggested over 50 years ago. However, lacking of chemically stable cathode material frustrated efforts to fabricate reliable cold cathode field emission devices [6]. Fortunately, carbon nanotube (CNT) was reported in 1991 [7] and it was suggested that this new material had great potential for use as an electron field emitter [8].

CNT has the inherent advantage of stable emission, long lifetimes, and low emission threshold potentials [9]. Lu et al. used CNT as cathode materials in X-ray source to replace conventional filament in 2002 for the first time [10]. The devices consisted of a multiple sources array within a single source [11,12]. Recently, they transferred their attention to radiation using CNT based X-ray source [13,14]. Compared with traditional X-ray source, CNT based X-ray source demonstrates vast advantages. Above all, generating little heat inside the cold cathode source is crucial for the minimization of an X-ray source and prolongs its lifetime. Moreover, the amount of electron emission is up to its surface electric field intensity, consequently CNT based X-ray source is readily

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* Corresponding authors.
E-mail addresses: yc.zhu@siat.ac.cn (Y. Zhu), yq.xie@siat.ac.cn (Y. Xie).

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controllable in a pulse operation by adjusting the grid voltage when the cathode is earthed or controlling the insulated gate bipolar transistor (IGBT) open and close with a designed pulse [12,15]. In addition, due to the desired gaussian distribution, the intensity distribution of the focal spot obtained from a pinhole measurement can provide better spatial resolution than either uniform or bimodal intensity distributions, not the double-peaked distribution commonly found on thermionic X-ray sources [16].

In view of the excellent properties of CNT materials, several groups have recently disclosed the concept of multi-source X-ray imaging using CNT based X-ray source. Jonathan S. Maltz et al. proposed a stationary-gantry tomosynthesis system for on-line image guidance in radiation therapy base on a 52-source cold cathode based X-ray source [6]. Zhou et al. proposed a spatially distributed multi-beam field emission X-ray source for stationary digital breast tomosynthesis [17–21]. The systems proposed by Jonathan S. Maltz and Zhou can obtain pretty high temporal resolution on account of no any mechanical movement during imaging process. But digital tomosynthesis only acquired fewer projections as compared to conventional CT scanners, it yielded images similar to conventional tomography with a limited depth of field [22]. Li et al. designed a multi-source instant CT for superfast imaging [23], but their system still needed several rotation times during covering the range of 360°.

It is always desirable to reduce the gantry motion and speed up scanning process. In this paper, a novel design of ultrafast micro-CT (UMCT) system is presented. It consists of 39 CNT based X-ray sources and 3 flat-panel detectors equally distributed in 2 concentric circles. When all the 39 sources have been activated according to a certain order, the gantry will be contrarotated 40° twice and 39 more projections are further acquired after each rotation. In the end, an iterative reconstruction algorithm is employed due to limited number of projections [24].

The remaining of this paper is organized as follows. In Section 2, the CNT based X-ray source and the design of UMCT system are introduced in detail, followed by the introduction of an iterative reconstruction algorithm. Section 3 provides the simulation results of the proposed system design. Discussions and conclusions are in Sections 4 and 5, respectively.

2. Methods

2.1. Ultrafast micro-CT (UMCT)

This section presents a detailed system design. The system consists of CNT based X-ray source and derivation of the relationship among various parameters.

2.1.1. CNT based X-ray source

A schematic diagram of the CNT based X-ray source is illustrated in Fig. 1(A). The source consists of CNT based field-emission cathode, focus electrode, grid electrode and anode housed in a vacuum chamber at 10^6 *torr* base pressure with a beryllium window. When the grid voltage reaches a certain value, the electron is emitted in the free state, then X-ray is generated under the interaction between high-speed electron and anode target. The grid voltage depends on the structure of CNT material and the distance between cathode and grid electrode.

Our home-made CNT based X-ray source is manifested in Fig. 1 (B). The size of CNT material placed on cathode is $2 \times 5\ mm^2$, the distance between the grid electrode and cathode is 250 μm. The current is controlled by the voltage between grid electrode and cathode, while the X-ray photon energy is determined by the acceleration voltage between the anode and the cathode. The focus electrode is also used in the present design, the focal-spot size is controlled by the potential of the focus electrode.

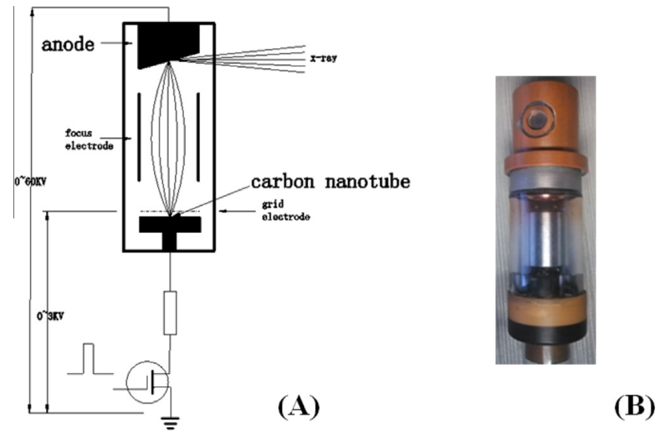

Fig. 1. CNT based X-ray source. (A) A schematic diagram of the CNT based X-ray source and (B) our home-made CNT based X-ray source.

2.1.2. System design

In order to achieve *in vivo* high—temporal—resolution image, an UMCT system is proposed as manifested in Fig. 2, which consists of 3 subsystems equally distributed in 2 concentric circles A and B. For each subsystem, there are n ($n \in N^+$) CNT based X-ray sources equally placed in the range of certain degree θ ($\theta \in (0, 2\pi)$) in the circle A of radius R ($R \in R^+$) and a flat-panel detector with the minimum size $d \times d$ ($d \in R^+$) facing the middle source of the corresponding n X-ray sources, which is placed in the circle B of radius l ($l \in R^+$). r ($r \in R^+$) is the radius of FOV.

In this design, since the overall size of the gantry geometry is limited in practice, the gantry configuration need to be carefully calculated following some principles: (1) θ must be a divisor of 120, like 40, for 360°—scanning after a limited number of rotation. (2) The detector must be placed in the appropriate position in order to acquire complete projection data. For example, the upper

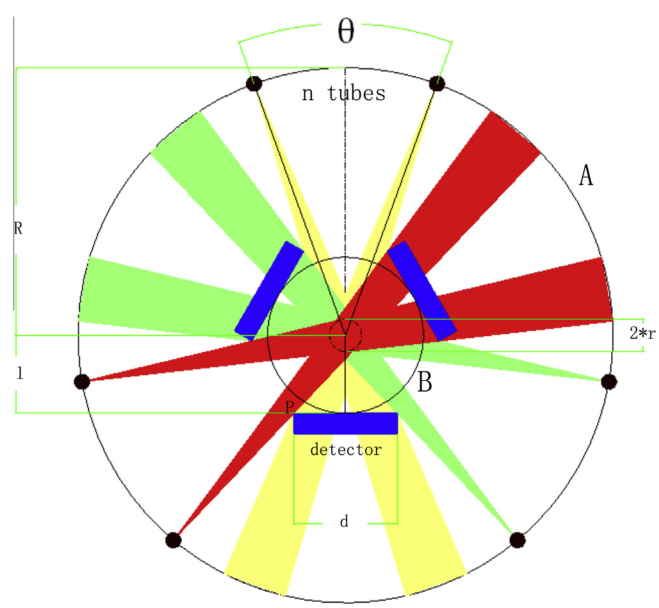

Fig. 2. 2D schematic diagram of UMCT. where θ is the covered range of a set of sources with n carbon nanotube based X-ray sources, R is the radius of circle A, r is the radius of FOV, l is the distance between the center of circle A and the flat-panel detector, d is the minimum size of flat-panel detector.

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