Medical Engineering and Physics 38 (2016) 1449–1457

Contents lists available at ScienceDirect

Medical Engineering and Physics

journal homepage: www.elsevier.com/locate/medengphy

Cone-beam micro computed tomography dedicated to the breast

Antonio Sarno[a,b,*], Giovanni Mettivier[a,b], Francesca Di Lillo[a,b], Mario Cesarelli[c], Paolo Bifulco[c], Paolo Russo[a,b]

[a] Dipartimento di Fisica "Ettore Pancini", Università di Napoli Federico II, Complesso Universitario di Monte Sant' Angelo, Via Cintia, snc, I-80126 Napoli, Italy
[b] Istituto Nazionale di Fisica Nucleare, Sezione di Napoli, I-80126 Napoli, Italy
[c] Dipartimento di Ingegneria Elettrica e Tecnologie dell'Informazione, Università di Napoli Federico II, I-80125 Napoli, Italy

ARTICLE INFO

Article history:
Received 10 March 2016
Revised 23 August 2016
Accepted 23 September 2016

Keyword:
Breast cancer
Diagnosis
Micro computed tomography
Phase contrast
MTF

ABSTRACT

We developed a scanner for micro computed tomography dedicated to the breast (BμCT) with a high resolution flat-panel detector and a microfocus X-ray tube. We evaluated the system spatial resolution via the 3D modulation transfer function (MTF). In addition to conventional absorption-based X-ray imaging, such a prototype showed capabilities for propagation-based phase-contrast and related edge enhancement effects in 3D imaging. The system limiting spatial resolution is 6.2 mm^{-1} (MTF at 10%) in the vertical direction and 3.8 mm^{-1} in the radial direction, values which compare favorably with the spatial resolution reached by mini focus breast CT scanners of other groups. The BμCT scanner was able to detect both microcalcification clusters and masses in an anthropomorphic breast phantom at a dose comparable to that of two-view mammography. The use of a breast holder is proposed in order to have 1–2 min long scan times without breast motion artifacts.

© 2016 IPEM. Published by Elsevier Ltd. All rights reserved.

1. Introduction

Computed tomography (CT) dedicated to the breast (breast CT, BCT) is an X-ray imaging technique that produces 3D images of the breast anatomy for breast cancer diagnosis. It features an isotropic resolution and absence of breast compression, the latter causing discomfort to the patients in mammographic exams as well as in digital breast tomosynthesis scans [1–3]. A recent review was provided by Sarno et al. [2]. BCT, introduced in 2001 [1,3] is able to overcome the well-known limits of mammography due to the superposition of the tissues, which may hide tumor masses and hence may reduce sensitivity.

In a BCT exam, the patient, lying on a table in prone position, inserts her breast in a hole in the bed and the breast, hanging freely from the hole, is imaged via an X-ray tube and a flat panel detector: the gantry rotates under the table in the coronal plane [1–3]. X-ray tube voltages from 49 to 80 kVp are routinely adopted in BCT setups. The prototype developed at University of Rochester and produced by a spinoff company [3–5] received FDA approval for diagnostic imaging of the breast in conjunction with standard two-view mammography [6], as well as the CE mark. The dose delivered to the breast during the imaging session via such a commercial scanner was estimated to be from 1.4 to 7.2 times higher than that used in 2-views mammography exams for the same breast [7].

A limit of first-generation BCT scanners is their relatively low spatial resolution, which ranges between 1.7 mm^{-1} and 5.6 mm^{-1}, when evaluated as the spatial frequency at which the modulation transfer function (MTF) curve reaches 10% ($MTF_{0.1}$) [1]. For comparison, the limiting resolution of full-field digital mammography (FFDM) reaches 12 mm^{-1} [8]. This relatively low spatial resolution might present problems for the BCT performance in detecting microcalcification clusters. In clinical studies, the BCT systems developed at University of Rochester [5] and at University of California Davis [9] detected 84.8% and 66.7% of the microcalcification clusters embodied in the imaged breast, respectively. Several factors limit the maximum BCT system spatial resolution: (i) the penumbra caused by the large size of the focal spot of the X-ray source (typically in the order of 0.3–0.4 mm nominal), (ii) the relatively large flat-panel detector pixel pitch (typically about 0.2 mm), (iii) the blurring introduced by the gantry motion, and (iv) the scintillator layer of the flat panel detector, usually thicker than that adopted in FFDM, for the need to detect more penetrating photons than in mammography. Many efforts have been made to reduce the influence of these limiting factors. The group at the University of Rochester developed a BCT scanner with a pulsed X-ray source in order to reduce gantry motion influence on the system spatial

* Corresponding author at: Dipartimento di Fisica "Ettore Pancini", Università di Napoli Federico II, Complesso Universitario di Monte Sant' Angelo, Via Cintia, snc, I-80126 Napoli, Italy.
E-mail address: sarno@na.infn.it (A. Sarno).

http://dx.doi.org/10.1016/j.medengphy.2016.09.012
1350-4533/© 2016 IPEM. Published by Elsevier Ltd. All rights reserved.

resolution, which reached 1.9 mm^{-1} [4]. The group at UC Davis improved the spatial resolution of its BCT prototype by embodying in the setup a pulsed X-ray tube and a high resolution CMOS flat panel detector with a pixel pitch of 75 µm (which works with an effective pixel pitch of 150 µm) coupled to a thin CsI(Tl) scintillator layer [10]. Thanks to these upgrades, the limiting spatial resolution of the first two prototypes, lower than 1.8 mm^{-1} [11], increased up to 3.6 mm^{-1} for the fourth prototype at UC Davis [12]. The high-resolution, direct-conversion, CdTe photon counting detector allowed the spiral BCT scanner developed at Erlangen University to reach a spatial resolution up to 5.3 mm^{-1} [13]. This gives the potential of showing microcalcifications with a diameter down to 150 µm [14,15].

Our group at the Department of Physics at University of Naples Federico II developed a first scanner for micro CT dedicated to the breast (BµCT). It embodied a high resolution CMOS flat panel detector with a pixel pitch of 50 µm and a 150-µm thick CsI(Tl) scintillator layer, and an X-ray source with a micro focal spot of 40 µm nominal size [16]. That prototype reached a limiting spatial resolution of 3 mm^{-1} [16]. The effective detector pixel pitch, the focal spot size and the spatial resolution of the prototypes developed at UC Davis, U Rochester, U Erlangen and U Naples are summarized in Table A1 (in Appendix). This work presents the characterization of the new BµCT scanner developed at U Naples (third prototype, Table A1): here, the X-ray tube has been replaced with a micro focal spot (7 µm) tube. In this work, the system MTF is described and the 1D noise power spectra (NPS) have been evaluated.

When an X-ray e.m. wave passes through an object, both its amplitude and phase change. Using phase-detection techniques, the spatially-resolved phase shift of the X-ray wave can be detected in a plane transverse to the beam axis. Soft tissues, which present small differences in attenuation coefficients as in the case of certain types of breast lesions and healthy tissues, can present larger differences in the emerging X-ray beam phase shift. Propagation-based phase contrast imaging (PB-PhC) is an imaging technique that does not need special optical elements in the beam path, apart from an X-ray incident beam with some degree of spatial coherence and a sufficient distance between the imaged object and the detector. It can be implemented either with synchrotron radiation sources [17,18,19] or in the laboratory using compact micro-focus X-ray tubes [20]. The advantages of the edge enhancement effects in projected images in PB-PhC [21,22] – produced by X-ray refraction at the boundary of different tissue structures – have already been shown in FFDM [23,24]. The micro-focal spot of the new scanner developed at U Naples, together with a sufficient distance between source and imaged object, produces an X-ray beam with sufficient spatial coherence, which makes this BµCT scanner a viable tool for PB-PhC for future clinical applications. In this work, the edge enhancement produced by the phase effects, both in planar and in 3D reconstructed images, was investigated.

A phantom study was conducted in order to test the capability of the BµCT scanner in showing phantom microcalcifications and soft tissue lesions with a dose similar to the one used in two-views FFDM and with a long scan time still considered adequate for in clinical use.

2. Materials and methods

2.1. The cone beam BµCT system

The cone-beam microCT scanner dedicated to the breast (Fig. 1) is a modular benchtop system, which embodies the components described below.

(1) A 75-W microfocus X-ray tube (Hamamatsu model L8121-03) with a selectable focal spot whose size of 7, 20 or 50 µm. This

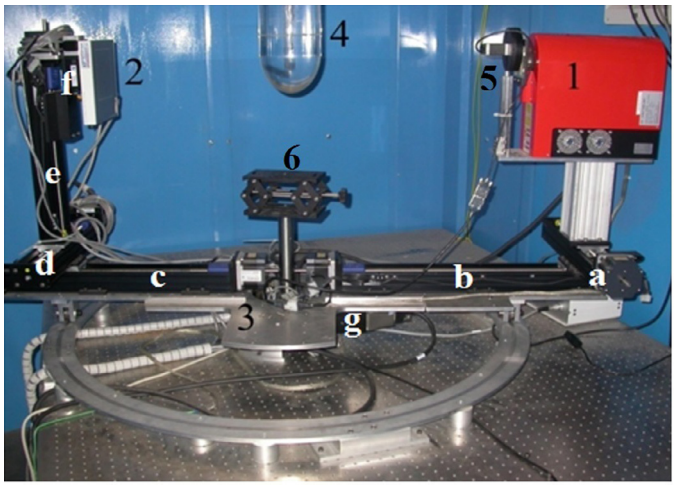

Fig. 1. Cone-beam microCT prototype scanner dedicated to breast imaging. (1) Microfocus X-ray tube and (2) CMOS flat panel detector, mounted on step-motor linear stages for variable-magnification imaging; (3) step-motor rotating gantry; (4) hanging PMMA breast phantom placed at isocenter and simulating a pendant breast; (5) rotating filter wheel for beam shutter; (6) post and lab jack at isocenter for hosting phantoms. Step-motor linear stages (a–e) and rotation stages (f, g) are used for setting the acquisition geometry and for gantry rotation. The system is mounted on an optical table (1.8 m × 1.2 m) and the patient bed can be positioned at 154 cm from the floor.

air-cooled tube has a fixed tungsten anode, a cone angle of 43° and it can be operated at a constant voltage between 40 kV and 150 kV; the output window is made of 0.2-mm Be and the added filtration is 1.58 mm Al. The measured half value layer (HVL) at 50 kV is 1.3 mm Al. The distance between source and isocenter is 612 mm. Fig. A1 (in Appendix) shows the tube output at isocenter as a function of tube voltage. Table 1 shows the scan time necessary to deliver a mean glandular dose (MGD) of 4 mGy to a 14-cm diameter and 50% glandular breast, calculated on the basis of the tube output at isocenter and on the basis of the monoenergetic normalized glandular dose (DgN) coefficients evaluated by Thacker and Glick [25]. The X-ray spectra were simulated with SpekCalc [26]. The duration of a full 360° scan is in the range of 69 s (@ 120 kV) to 336 s (@ 40 kV) when the focal spot is set to 7 µm, decreasing correspondingly to 11 – 161 s with a 50-µm focal spot. Hence, scans in which the patient is holding her breath are not feasible, and patient specific tools capable of immobilizing the breast during the scan are necessary.

(2) A CMOS flat panel detector (Hamamatsu mod. C7942CA-02) with a 150-µm thick CsI:Tl scintillator layer, with a sensitive area of 12 cm × 12 cm and a 50-µm pixel pitch. According to specifications, the C7942CA detector has a spatial resolution of 8 lp/mm (at 5% Contrast Transfer Function); the measured MTF reaches 10% at 6 mm^{-1} [16]. The detective quantum efficiency of the detector is higher than 0.4 at zero frequency (at 60 kVp, mean energy = 36.3 keV) [27]. The frame rate is 2 fps at 1 × 1 pixel binning (2.13 fps measured rate), 4 fps at 2 × 2 binning and 9 fps at 4 × 4 binning. The analog-to-digital conversion produces 12 bit/pixel signals in analog to digital units (ADU). Fig. A2a (in Appendix) shows the results of a test for assessing the detector lag, confirming the limited relevance of this effect for the CMOS flat panel detector.

We acquired four hundred consecutive frames and then calculated the average pixel value in a 400 × 400 pixels region of interest (ROI). The curves for 50, 80 and 100 kV were normalized to their starting values. The results showed a weak upward trend where the mean signal increases up to 0.7% (at 80 kV) from its starting value after 400 consecutive projections. Previous studies [28] showed an asymptotic trend for this curve. Fig. A2b (in Appendix) shows linearity of detector signal (evaluated as the mean

Download English Version:

https://daneshyari.com/en/article/5032775

Download Persian Version:

https://daneshyari.com/article/5032775

Daneshyari.com