FOCUSED TOMOGRAPHY

BACKGROUND

[0001] The National Institutes of Health (NIH) has noted that Computerized tomography (CT) use is up over 2000% in the last 20 years. In the 1990's, CT machines were primarily used for post-diagnostic tests. They are now often used for prediagnostic tests especially in the case of trauma such as car accidents, etc. Accordingly, CT use has become so prevalent that the cumulative radiation dosage which patients are exposed to during their lifetime has increased dramatically.

BRIEF DESCRIPTION OF THE DRAWINGS

[0002] Many aspects of the present disclosure can be better understood with reference to the following drawings. The components in the drawings are not necessarily to scale, emphasis instead being placed upon clearly illustrating the principles of the present disclosure. Moreover, in the drawings, like reference numerals designate corresponding parts throughout the several views.

[0003] FIG. 1 is a diagram illustrating the basics of the projection angles, coordinates, and vectors, $r, \theta, \vec{\theta},$ and $\vec{x}$ in accordance with the present disclosure.

[0004] FIG. 2 is a diagram depicting a region of interest, local data, and non-local data in accordance with embodiments of the present disclosure.

[0005] FIGS. 3A-3B show focused tomography images on a portion of a subject's spine and hip in accordance with embodiments of the present disclosure.

[0006] FIG. 4 is a diagram showing the correction term necessary for the inverse Fourier transform in polar coordinates in accordance with embodiments of the present disclosure.

[0007] FIG. 5 is a diagram showing kernel decompositions $k_l(r)$ and $k_h(r)$ in accordance with embodiments of the present disclosure.

[0008] FIGS. 6A-6B are diagrams illustrating a standard Radon transform and sampling schemes which have been suggested before. Figure 6C shows the sampling scheme in accordance with embodiments of the present disclosure. High frequency on the central portion, and low frequency off the central portion.

[0009] FIGS. 7A-7F are diagrams illustrating image reconstructions for a central region and a non-central region from full and reduced data sets in accordance with embodiments of the present disclosure.

[0010] FIGS. 8A-8B are diagrams comparing image reconstructions from local data, in which FIG. 8A illustrates localized reconstruction in accordance with embodiments of the present disclosure and FIG. 8B illustrates an image reconstruction from totally local data.

[0011] FIG. 9 is a diagram of a side view of an exemplary CT scanner showing the location of the proposed sliding collimators in relation to other key components of an exemplary CT system in accordance with embodiments of the present disclosure.

[0012] FIGS. 10A-10B are diagrams illustrating views of the sliding collimators looking up from underneath the x-ray tube/transmitter, showing (A) the collimators fully open and (B) the collimators fully closed in accordance with embodiments of the present disclosure.

[0013] FIGS. 11A-11B are diagram illustrating views of the sliding collimators looking up from underneath the x-ray tube at two possible configurations in accordance with embodiments of the present disclosure.

[0014] FIGS. 12A-12D are diagrams showing sample anteroposterior and lateral topograms with lines indicating the anatomy to be included in a focused CT in accordance with the present disclosure. In these examples, the exams are for (A-B) right shoulder and (C-D) right hip of a human subject.

[0015] FIGS. 13A-13D are diagrams showing side views of an exemplary CT scanner showing how the sliding collimators shift the focused scan field-of-view during the rotation of the x-ray tube and detector in accordance with embodiments of the present disclosure.

DETAILED DESCRIPTION

[0016] The present disclosure describes systems and methods for improving the focusing of computerized tomography (CT) machines, such as all of those currently in use. This will allow one to image portions of the human anatomy, such as the spine, shoulders, hips, and elbows, without exposing an entire slice of the human body to a full radiation (x-ray) dose.

[0017] Computerized tomography (CT) has become a standard diagnostic tool in modern medicine. In the last 30 years, however, CT use has become so prevalent that the cumulative radiation dosage which patients are exposed to during their lifetime has increased dramatically. The risk/reward tradeoffs for these diagnostic tools are hard to quantify. Often times a physician is only interested in a limited region of interest (ROI) which has been identified through clinical means or previous imaging. We believe that we can reduce the risk when a limited ROI is of interest by significantly reducing the total radiation dosage. The reward of this imaging will be preserved, since the images of the ROI will be nearly identical to those achieved with full radiation dosage.

[0018] The original investigations into region of interest (ROI) tomography, called local tomography, did not return the actual density of the ROI. Rather these returned an altered or transformed image which did preserve edges between varying tissues. The goal of local tomography is returning the actual image of the ROI. This is possible when one realizes that the low frequency components of the image are the only components which need non-local x-rays for their estimation. The high frequency components can be measured with the local line integral measurements which pass through the ROI. Thus, one does not need to send high radiation on paths which do not intersect the ROI to find the high frequency components of the image. In FIG. 2, an illustration of local versus non-local x-ray paths is depicted. Low radiation measurements can then measure the non-local information.

[0019] In the figure of FIG. 2, concepts are depicted for a region of interest (ROI), local data, and non-local data. The region of interest (ROI), is the dark portion, which might be an approximation for the spinal column of a patient. A portion of the lines represents x-ray paths which pass through the ROI, and are therefore called local x-ray paths. The other lines represent x-ray paths that do not intersect the ROI, and are therefore referred to as non-local x-ray paths. The systems and methods of the present disclosure will minimize radiation along non-local x-ray paths.

[0020] Thus to provide a complete and accurate reconstruction of the ROI, two earlier methods feature (a) regular sampling of the x-rays through the ROI in order to reconstruct the high frequency components of the image via a computer processor of the CT scanner, and (b) low radiation sampling of the x-rays which do not intersect the ROI, allowing for the recovery of low frequency components of the image. The combination of these methods results in an accurate reconstruction of the ROI with greatly reduced radiation dosages. These earlier methods work very well but only allow 0, 1 sampling, i.e. either an x-ray is measured along a line integral or it is not.

[0021] The present disclosure concentrates on new improved methods and systems, which allow for variable sampling of the x-rays by featuring regular dosage radiation through the ROI & dramatically reduced dosage radiation through the portions of the body outside of the ROI. The lower dosage measurements outside of the ROI will be sufficient because they will only be utilized to determine the low-frequency components of the ROI image.

[0022] Thus, the lower signal to noise ratio (SNR) measurements (at lower radiation dosages) can be averaged producing adequate SNR measurements for these low frequency components. The detailed high-frequency components of the image will have standard SNR measurements (at standard radiation dosages). Thus the image in the ROI can be made arbitrarily close to that using complete radiation dosage, with dramatically lower dosages.

[0023] Various systems and methods of the present disclosure will allow physicians to view and monitor regions of interest using radiation dosages outside the ROI which are 10% of the standard dosage. This work will augment, rather than compete with, efforts which primarily focus on improving receiver sensitivity.

[0024] Current clinical practice is to choose a body segment to be imaged, and expose the entire portion to the same radiation levels. Thus, if the right shoulder is the region of interest, the entire chest from the neck to below the shoulder will be exposed to the same amount of radiation. In accordance with teachings of the present disclosure, physicians will be enabled to ask for only an image of the right shoulder or ROI. Accordingly, systems and methods of the present disclosure may then concentrate radiation on that ROI and use a greatly reduced amount of radiation on the rest of the upper chest cavity and neck. Similarly, when doing spinal imaging, the radiation will be concentrated on that portion of the spine which is of interest, minimizing the radiation elsewhere, as demonstrated by FIG. 3A showing focused tomography on a portion of a subject’s spine in accordance with embodiments of the present disclosure. This development will constitute a shift in clinical practice, which can eliminate up to 90% of the radiation dosage.

[0025] This focused tomography may be especially meaningful concerning diagnostic endeavors with children and pregnant women. In this disclosure, we discuss the imaging of tissue structures in cylindrical volumes (while also contemplating the use of non-cylindrical volumes). The mathematical methods of the present disclosure promise the possibility of imaging within formed structures that can directly follow the outline of the anatomy under concern.

[0026] We are able to accomplish this focused tomography through a thorough understanding of the CT reconstruction process. CT data is generally collected in a helical fan-beam geometry. For the discussion of the reconstruction process, it is easier if we assume that this data has been reformatted into a parallel beam data set. Thus, we assume that the data can be viewed as a function of two variables, angle $\vec{\theta}$ and distance r from the origin, or that the data, which we refer to as the projections or sinogram, has the form

$$P_f(r,\vec{\theta}) = \int f(r\vec{\theta} + t\vec{\theta}^{\perp})dt.$$

The basics of the CT projection is illustrated in FIG. 1.

[0027] To this end, let us recall that the natural coordinates for tomography are $x = r\vec{\theta} + t\vec{\theta}^{\perp} = r$. Moreover, $\vec{x} \cdot \vec{\theta}^{\perp} = r$. We want to see what the Fourier coefficients propagating at a fixed direction $\vec{\theta}$ have in common with the projections, so we consider

$$\hat{f}(s\vec{\theta}) = \frac{1}{2\pi}\int P_t(r,\vec{\theta})e^{irs}dr. \qquad (1)$$

Thus, a central slice of the two dimensional Fourier transform of $f(x,y)$, i.e. $\hat{f}(s\vec{\theta})$ can be obtained from the one dimensional projections of the function or $P_f(r,\vec{\theta})$. Formally stated, we have Theorem 2.1 (Radon Transform, or Central Slice Theorem), as described below.

Theorem 2.1.

The one dimensional Fourier transform of $P_f(\vec{\theta},r)$ is given by the central slice of the two dimensional Fourier transform, or $\hat{f}(s\vec{\theta})$. Mathematically, $\mathcal{F}_1(P_f(r,\vec{\theta}) = \frac{1}{\sqrt{2\pi}}\hat{f}(s\vec{\theta})$.

From this formula one can quickly derive the Filtered Backprojection formula, which was the basis of the 1979 Nobel Prize.

$$f(\vec{x}) \approx \frac{1}{2\pi}\int_0^{\pi}\int_{-\infty}^{\infty}\hat{f}(s\vec{\theta})|s|w(s)\, e^{irs}dsd\theta \qquad (2)$$

$$= \int_0^{\pi} (P_f(r,\theta) * \mathcal{F}^{-1}(|s|w(s)))(\vec{x}\cdot\vec{\theta})d\theta.$$

By choosing w(s) appropriately, we can make the approximation above arbitrarily small. If we denote $\mathcal{F}^{-1}(|s|w(s)) = k(r)$ then we have

$$f(\vec{x}) \approx \int_0^{\pi} (P_f(r,\theta) * k(r)))(\vec{x}\cdot\vec{\theta})d\theta, \quad (3)$$

where * denotes convolution.

[0028] One problem with Equation (3) is that the kernel $k(r)$ is very broad as a function of r, and as a result, radiation measurements must be taken far from the region of interest. The reason for this kernel being broad is the jump discontinuity of the derivative of the function $|s|$ at the origin from -1 to 1. Recall that $|s|$ is the necessary term due to the polar coordinates used in the Fourier inversion of the filtered backprojection formula (2). The basic theorems of Fourier analysis dictate that this kernel cannot decay quickly.

[0029] We solve this problem by separating the discontinuity at the origin of $|s|$ into separate portions: $|s|w_2(s)$ at the origin and $|s|(1-w_2(s))$ away from the origin, as illustrated in FIG. 4. The corresponding inverse Fourier transforms will be a low frequency kernel $k_l(r)$ which is the inverse Fourier transform of $|s|w_2(s)$, and a low frequency kernel $k_h(r)$ which is the inverse Fourier transform of $|s|(1-w_2(s))$.

[0030] Referring back to FIG. 4, the figure shows the correction term necessary for the inverse Fourier transform in polar coordinates above, i.e. $|s|$. The problem with $|s|$ is the jump discontinuity at the origin of the derivative of $|s|$, from 1 to negative 1. This dictates that the inverse Fourier transform of $|s|w(s)$, even with a suitably smooth window $w(s)$, will be very wide, rather than narrow. This can be solved as above, with one “low frequency” term at the origin and a “high frequency” term. The corresponding decomposition of the kernel $k(r) = k_l(r) + k_h(r)$ will result in a low frequency kernel as a function of radius, $k_l(r)$, which is not locally supported, and a high frequency kernel $k_h(r)$ which is very narrow as a function of radius.

[0031] Thus, we have $k(r) = k_l(r) + k_h(r)$. The filtered backprojection algorithm now looks like

$$f(\vec{x}) \approx \int_0^\pi (P_f(r,\theta) * k(r)))(\vec{x} \cdot \vec{\theta}) d\theta$$

$$= \int_0^\pi (P_f(r,\theta) * k_l(r)))(\vec{x} \cdot \vec{\theta}) d\theta + \int_0^\pi (P_f(r,\theta) * k_h(r)))(\vec{x} \cdot \vec{\theta}) d\theta, \quad (5)$$

$$= f_l(\vec{x}) + f_h(\vec{x}), \quad (6)$$

and, we will reconstruct, via a computer processor, the low and high frequency terms of $f(\vec{x})$ separately. The kernels are illustrated in FIG. 5 in which we show the kernel decompositions $k_l(r)$ and $k_h(r)$.

[0032] The energy of $k_h(t)$ is contained within the interior 9 pixels of the current digitization or 9/512 to an accuracy of 1/10000. The energy concentration of $k_l(r)$, similarly measured, takes 175 terms. The low frequency terms take a great deal of non-local information, and the high frequency terms can be measured locally.

[0033] Initially, there seems to be no advantage to the change to two kernels $k_l(r)$ and $k_h(r)$ from a radiation reduction standpoint. The low frequency kernel will require the gathering of large quantities of data from outside the region of interest. Thus, there is no apparent win in the fact that we can calculate the high frequency component $f_h(\vec{x})$ from completely local measurements. One must understand the structure of the projections, and corresponding structure of the filtered backprojection algorithms to see how to solve our problems with the low frequency reconstruction $f_l(\vec{x})$. The structure theorem for the projections or Radon transform states that $P_f(\vec{\theta}, r) = (1 - r^2)^{-1/2} \sum_{l=0}^{\infty} T_l(r) h_l(\theta)$, where $T_l(r)$ are the Chebyshev polynomials. Taking the Fourier transform of this yields

$$\hat{f}(s\vec{\theta}) = \hat{P}_f(\theta, s) = (\tfrac{\pi}{2})^{-\frac{1}{2}} \sum_{l=0}^{\infty} i^{-1} J_l(s) h_l(\theta) \quad (7)$$

$$= (\tfrac{\pi}{2})^{-1/2} \sum_{l=0}^{N-1} i^{-1} J_l(s) h_l(\theta) + (\tfrac{\pi}{2})^{-\frac{1}{2}} \sum_{l=N}^{\infty} i^{-1} J_l(s) h_l(\theta) \quad (8)$$

$$= \hat{f}_l(s, \theta) + \hat{f}_h(s, \theta) \quad (9)$$

where $J_l(s)$ are the Bessel functions, and $h_l(\theta)$ is a trigonometric polynomial of order l. The key to understanding Equation (7) is that the low frequency terms in s, which are the Bessel functions, are only multiplied in frequency by the low order terms $h_l(\theta)$. Thus, we do not have to measure the low frequency terms for many angles θ in order to accurately determine the complete low frequency components of the image.

[0034] The sampling of the projections, or Radon transform, is illustrated in FIGS. 6A-6C, in which FIG. 6A illustrates a standard Radon transform or sinogram and FIG. 6B illustrates the sampling recommended for a central region of an image. While this center sampling technique greatly reduced the radiation levels by as much as 90%, this technique was designed for parallel beam geometries and is not completely feasible for fan beam geometries. In particular, for standard fan-beam CT machines, the sampling is 0-1, meaning that the x-ray tube would either have to be shut off or modulated very quickly to accomplish this type of sampling.

[0035] We will now outline the refinements and improvements for the present disclosure that will make this project feasible in a fan-beam geometry as well as a parallel beam geometry. Accordingly, FIG. 6C illustrates an exemplary sampling scheme which we are espousing for localized imaging in this disclosure. As opposed to the center sampling scheme (see FIG. 6B), an exemplary sampling technique in accordance with embodiments of the present disclosure (see FIG. 6C) is for an off-centered region. This off-centered type of sampling scheme will easily work with fan-beam geometries.

[0036] As described previously, earlier methods concentrated on 0-1 sampling. Namely, a linear x-ray beam would be either sampled or not sampled. We will relax this condition, and work to find the optimal solution to minimize radiation. Thus, we will either sample at the necessary high-dosage rate, which is required for appropriate SNR and resolution in the ROI, or a variable lower-dosage rate, which is all that is necessary outside the ROI.

[0037] This optimal solution to this problem is necessarily better than the optimal solution to the 0-1 sampling problem. Any time more variables are added to an optimization problem, the solution necessarily gets better. Moreover, this will allow us to design the system in a way that is easily implementable in hardware. In one embodiment, we can use static or non-adaptive/dynamic filters in front of the x-ray transmitter, to alter the beam for the appropriate reduced dosage.

[0038] We will now illustrate this approach, both mathematically and visually. Assume for now that the region of interest is circular (with radius r_1) and that the x-ray scanner is centered on the center of the circle when θ = 0. Then, there is a distance *d* from the global center of the scanner to the center of the ROI. As the x-ray transmitter moves with θ, the center of the circle will then be a distance $d(\theta) = dsin(theta)$ off of the center of the gantry. Thus we want to gather a full data set of the x-rays which pass through the ROI, which represent our first data set $P_f^h(\theta,r) = P_f(\theta,r)$ where $0 \leq \theta \leq \pi$ and $r \in [dsin(\theta) - r_1,\ dsin(\theta) + r_1]$. This first data set is gathered with full radiation dosage, just as if you were going to image the whole slice. Therefore, it will have a relatively high SNR. The notation $P_f^h(\theta,r)$ recognizes that this data will be used to reconstruct the high frequency details of the image.

[0039] A second data set is then gathered from all of the lines or projections which did not intersect the region of interest. The second data set is our low frequency data set $P_f^l(\theta,r) = P_f(\theta,r)$, where $r \notin [dsin(\theta) - r_1,\ dsin(\theta) + r_1]$. This low frequency data set is gathered with minimal radiation, will have very low SNR, and will only be needed to reconstruct the low frequency portion of the image and will not affect the final image in the ROI.

[0040] We now simply combine the data sets to get our approximate, noisy sinogram or Radon transform $P_f(\theta,r) \approx P_f^h(\theta,r) + P_f^l(\theta,r)$, noting that we have sampled all of the Radon transform, some of it at high SNR and some at low SNR. Our final reconstruction will be

$$f(\vec{x}) \approx \int_0^{\pi} (P_f(r,\theta) * k(r)))(\vec{x}\cdot\vec{\theta})d\theta, \tag{10}$$

$$= \int_0^{\pi} (P_f(r,\theta) * k_h(r)))(\vec{x}\cdot\vec{\theta})d\theta + \int_0^{\pi} (P_f(r,\theta) * k_l(r)))(\vec{x}\cdot\vec{\theta})d\theta$$

$$= \int_0^{\pi} (P_f^h(r,\theta) * k_h(r) + P_f^l(r,\theta) * k_h(r))(\vec{x}\cdot\vec{\theta})d\theta$$

$$+ \int_0^{\pi} (P_f^h(r,\theta) * k_l(r) + P_f^l(r,\theta) * k_l(r)))(\vec{x}\cdot\vec{\theta})d\theta$$

$P_f^h(r,\theta) * k_h(r)$ is the term which will yield most of our high resolution image and is well sampled through the ROI. This is the foundation of our reconstruction. The second term $P_f^l(r,\theta) * k_h(r)$ will be essentially zero for any contribution to the ROI, since $P_f^l(r,\theta)$ is only sampled for lines that do not intersect the ROI, and $k_h(r)$ is extremely well localized as we commented in FIG. 4. The term $P_f^h(r,\theta) * k_l(r)$ will yield a high partial estimate for the low frequency components, while $P_f^l(r,\theta) * k_l(r)$ gives a rather low SNR estimate for the low frequency components.

[0041] Thus, we have a very high SNR estimate for the high frequency components inside the ROI. We must recall the structure of the Radon transform to realize why the low frequency component is not affected by the low SNR estimates. Recall from Equation (7) that the low frequency components are only affected by low frequency sines and cosines with respect to θ or the Fourier transform

$$f_l(s\vec{\theta}) = \sum_{l=0}^{N-1} l^{-1} J_l(s) h_l(\theta).$$

Therefore, since we are estimating very few parameters in the low frequency component and have a great number of data samples, the law of large numbers will yield a very solid estimate for the low frequency component.

[0042] This process is illustrated in FIGS. 7A-7F, in which FIGS. 7A-7B illustrate the radiation exposure recommended for a central region and a non-central region of an image. The original non-noise related reconstruction is shown at FIG. 7A and the noisy reconstruction is shown at FIG. 7B. As opposed to the 0-1 sampling scheme, the adaptive sampling method in accordance with embodiments of the present disclosure is designed for arbitrary geometries, and can be implemented in a fan-beam geometry.

Accordingly, FIGS. 7C-7D show the ROI reconstructed with a full data set (in FIG. 7C), and with the reduced data set (in FIG. 7D) in accordance with embodiments of the present disclosure. Additionally, FIGS. 7E-7F show an off center ROI reconstructed from a large data set (in FIG. 7E) and small data sets (in FIG. 7F) in accordance with embodiments of the present disclosure. The radiation reduction was 92% for FIGS. 7D and 7F. A natural question might be “Do we really need to process any data which doesn’t pass through the ROI?”

[0043] To address this inquiry, FIGS. 8A-8B are diagrams illustrating a reconstruction using nearly-local data and completely local data. FIG. 8A is the localized reconstruction as was shown in FIGS. 7A-7F using nearly-local data. FIG. 8B is the reconstruction with totally local data, ignoring the steps for the localized data (nearly-local) as we outlined above. From a review of FIG. 8B, it is apparent that the quality of image reconstruction for FIG. 8B is not on par with the quality of the image reconstruction for FIG. 8A.

[0044] In order to test the hypothesis that local tomography can provide equal resolution and detail while greatly reducing radiation exposure, two data sets, one with low radiation dosage and one with high radiation dosage, can be acquired using appropriate testing phantoms. For this test, we can use high fidelity phantoms fabricated to mimic critical anatomical regions and use radiation-tissue equivalent materials. The spatial frequencies in these phantoms can match or be more severe than those in clinical practice. For optimal filter designs, we envisage high dosage within the ROI, and a dosage which falls off quickly when the lines do not pass through the ROI. The high dosage will be used for local high frequency imaging and the low dosage for whole section imaging of low frequency details.

[0045] In addition to artificial phantoms, testing can also utilize reasonably sized anesthetized pigs (50 lbs). We can anesthetize our subject pigs with an appropriate amount of Xylazine and Ketamine. Pigs obviously have similar anatomy to humans and are very inexpensive for the purposes of testing. This can eliminate most questions about the biological validity of our studies without the need for human studies.

[0046] To assess the comparative image quality of the two methods (i.e. full dosage ROI scanning and reduced dosage ROI scanning), we can utilize various metrics. The obvious first metric can be simple mathematical mean square error measurements. These measurements do not, however, generally yield a true measure of the image quality. Therefore, we can utilize visual quality metrics, which have been developed in Computer Vision over the past 15 years to assess the relative value of both methods. Finally, we can enlist a significant number of qualified Physicians, Surgeons, and Radiologists to view this work. We can make the study double blind, by showing images within the ROI from both the full exposure and reduced exposure methods.

[0047] In various embodiments, static filters can be developed and used in conjunction with our scanners. These will focus the radiation on the ROI, and eliminate unnecessary non-local radiation. These static filters will only be able to gather data on a centralized ROI. We will test this by repositioning the phantoms, or pig subjects on the central axis of the CT machine. In various embodiments, adaptive filters may be used as alternatives to static filters.

[0048] Additionally, in various embodiments, designed filters will be active, gathering data through a non-centered ROI at near real-time gantry speeds. Cramer-Rao statistical bounds will be attempted to prove that we are using minimal data sets. Also, in various embodiments, designed filters will be fully automated, so that a physician can choose his ROI, and have the machine optimally reduce (e.g., via a controller device) the radiation dose in accordance with the present disclosure.

[0049] Accordingly, in one embodiment, a designed filter will be a static filter, in which we will look to build a filter, or filters, which will alter the output of the CT machine producing a desirable localized radiation profile. This is possible with a number of materials, such as Aluminum, by merely adjusting the depth of the filter. A stable material which will not degrade under the radiation exposure and which can be made thin enough to not interfere with the gantry is used in certain embodiments.

[0050] In one embodiment, a designed filter will be a mechanically active, materially static filter. Accordingly, after we have a static filter design, we will look to mechanically move it, keeping the focus of the radiation on the region of interest.

[0051] In the present disclosure, we consider cylindrical ROIs. This is consistent with the spine and shoulders, but perhaps not optimal for hip imaging. However, we contemplate that the mechanically-active filter would be capable of essentially imaging any ROI, including non-cylindrical ROIs. The radiation reduction off an exemplary cylindrical system is very substantial. We do not imagine decreasing this by more than 2 or 3 times with more advanced methods.

[0052] The foregoing describes the sequence of steps required to perform the novel method of focused computed tomography (CT) in one embodiment. In general, a computed tomography scanner features a ring or cylinder for a gantry, in which a subject is positioned. An x-ray tube and an x-ray detector are positioned opposite of each other and rotate around the gantry as x-ray images are acquired. A body scanning filter, such as bowtie filter, is generally positioned in front of the x-ray tube to shape the x-ray beam and reduce the range of x-ray energies that reach the subject, such as reducing the beam intensity at the periphery of the x-ray beam that is transmitted. Additionally, a static pre-patient collimator may be positioned between the filter and the patient. In accordance with the present disclosure, an additional adaptive collimator device can also be positioned between the pre-patient collimator and the patient.

[0053] For example, an exemplary CT scanner of the present disclosure provides two sliding collimators at the location indicated in FIG. 9 which is between an existing pre-patient collimator and the opening of the gantry where a patient or subject would be positioned. The collimators slide back and forth using motors, to permit the opening between them to be adjustable in size from fully open (i.e., no restriction to the scan field-of-view (SFOV)) to fully closed, as shown in FIGS. 10A-10B. The collimator opening may be adjusted to any size between these extremes by a controller device coupled to the motors and/or actuators for the sliding collimators.

[0054] In one embodiment, the position of the opening is adjustable laterally to create a focused field-of-view anywhere within the scan field-of-view, as shown in FIGS. 11A-11B. The figures show views of the sliding collimators looking up from underneath the x-ray tube. FIGS. 11A and 11B show two possible configurations of the size and position of the opening, in which the installed collimators will be able to produce an opening at any lateral position within the SFOV and with any size opening. Accordingly, the collimator movement can occur quickly enough to adjust the size and position of the opening during each rotation of the x-ray tube around the gantry.

[0055] In one embodiment, the sliding collimators are made of a material with sufficient thickness and density to reduce the measured air kerma to approximately one-tenth of the air kerma that would be measured in the beam exiting the bowtie filter and any other pre-patient filter. The exact thickness of the collimator depends on the beam quality of the CT scanner, which is dependent on manufacturer and model.

[0056] For structural integrity and the ability to adjust position quickly, the material can be a hard metal that can achieve this attenuation with a thickness of no more than a few millimeters. Appropriate materials that meet the requirements of sufficient thickness/density and structural integrity include but are not limited to copper or tungsten, in various embodiments. Approximate thicknesses would be about 6 mm of copper or 3 mm of tungsten, in various embodiments.

[0057] In one embodiment of an exemplary scanning procedure, the patient is placed on the table and positioned by the CT technologist for the CT scan. No special positioning of the patient specific to the application of a focused CT is required. The technologist acquires anteroposterior (AP) and lateral topograms. The anatomy to be included in the focused CT is marked on both topograms (in FIGS. 12A-12D for (A-B) right shoulder and (C-D) right hip of a human subject). An exemplary CT scanner is configured to process this information to calculate the size and position of the opening in the sliding collimators in accordance with embodiments of the present disclosure.

[0058] The acquisition parameters such as kV, mA, rotation time, and pitch should be the same as those that would be used in clinical exams without the sliding collimators. No adjustment to these techniques is required. Systems using mA modulation calculated from the topograms should continue to function correctly with the sliding collimators in place. Systems calculating mA modulation on the fly will likely need to be switched to a manual mA, in various embodiments.

[0059] In various embodiments, the sliding collimators adjust continuously or repeatedly during the CT acquisition to restrict the SFOV to the area of interest (as shown in FIGS. 13A-13D). The tissues outside the SFOV receive only the primary radiation that passes through the material of the sliding collimator, which is about 10% of the exposure within the area of interest. In FIGS. 13A-13D, four positions during CT acquisition are shown, but the collimator movement may be continuous throughout the rotation in accordance with embodiments of the present disclosure.

[0060] Functionality of an exemplary CT scanner in certain embodiments of the present disclosure or portions thereof can be implemented in hardware, software, firmware, or a combination thereof. Such software or firmware can be stored in a computer readable medium, such as memory and be executed by a suitable instruction execution system. If implemented in hardware, the hardware can be implemented with any or a combination of the following technologies, which are all well known in the art: a discrete logic circuit(s) having logic gates for implementing logic functions upon data signals, an application specific integrated circuit (ASIC) having appropriate combinational logic gates, a programmable gate array(s) (PGA), a field programmable gate array (FPGA), *etc.*

[0061] In the context of this document, a "computer-readable medium" can be any means that can contain, store, communicate, or transport a program for use by or in connection with an instruction execution system, apparatus, or device. The computer readable medium can be, for example but not limited to, an electronic, magnetic, optical, electromagnetic, infrared, or semiconductor system, apparatus, or device. More specific examples (a nonexhaustive list) of the computer-readable medium would include the following: an electrical connection (electronic) having one or more wires, a portable computer diskette (magnetic), a random access memory (RAM) (electronic), a read-only memory (ROM) (electronic), an erasable programmable read-only memory (EPROM or Flash memory) (electronic), an optical fiber (optical), and a portable compact disc read-only memory (CDROM) (optical).

[0062] It should be emphasized that the above-described embodiments are merely possible examples of implementations, merely set forth for a clear understanding of the principles of the present disclosure. Many variations and modifications may be made to the above-described embodiment(s) without departing substantially from the principles of the present disclosure. All such modifications and variations are intended to be included herein within the scope of this disclosure.

CLAIMS

Therefore, at least the following is claimed:

1. A focused tomography system comprising:

an x-ray transmitter that is configured to emit a radiation beam;

an x-ray detector that is configured to detect incident radiation from the radiation beam; and

an adaptive collimator device arranged between the x-ray transmitter and the x-ray detector, wherein the adaptive collimator device provides a pair of sliding collimators that actuate to adjust a size of an opening between the sliding collimators to be adjustable laterally in size, wherein the adaptive collimator device is movable, via one or more motors, to position the opening at any lateral position within a scan field-of-view and to configure a size of the opening within a range from fully opened to fully closed during each rotation of the x-ray transmitter around a subject.

2. The system of claim 1, further comprising a controller device connected to the x-ray transmitter that is configured to cause the x-ray transmitter to emit the radiation beam at a first radiation level when a path of the radiation beam intersects a region of interest of the subject and cause the x-ray transmitter to emit the radiation beam at a second radiation level when the path of the radiation beam does not intersect the region of interest of the subject, the second radiation level being less than the first radiation level.

3. The system of claim 2, further comprising a pre-patient filter arranged between the x-ray transmitter and the adaptive collimator device, wherein the pre-patient filter comprises a mechanical filter that is controllable by the controller device to reduce the radiation level and focus radiation from the x-ray transmitter on the region of interest.

4. The system of claim 1, wherein the collimator device is formed from Aluminum, Copper, or Tungsten material.

5. The system of claim 1, further comprising a pre-patient filter arranged between the x-ray transmitter and the adaptive collimator device, wherein the pre-patient filter comprises a static filter that focuses radiation from the x-ray transmitter on a region of interest.

6. A focused tomography method comprising:

arranging an adaptive collimator device between an x-ray transmitter and an x-ray detector of a gantry for a computerized tomography scanner, wherein the adaptive collimator device provides a pair of sliding collimators;

adjusting, via a controller device of the computerized tomography scanner, a size of an opening between the sliding collimators in size in order to restrict a scan field-of-view to a region of interest; and moving the adaptive collimator device to position the opening at any lateral position within the scan field-of-view.

7. The method of claim 6, further comprising emitting a radiation beam at a first radiation level when a path of the radiation beam intersects the region of interest and emitting the radiation beam at a second radiation level when the path of the radiation beam does not intersect the region of interest, the second radiation level being less than the first radiation level.

8. The method of claim 7, further comprising:

detecting, via an x-ray detector, incident radiation from the radiation beam at the first radiation level;

detecting, via the x-ray detector, incident radiation from the radiation beam at the second radiation level; and

reconstructing, via a processor, an image of the region of interest, wherein high frequency components of the image are determined solely from the incident radiation at the first radiation level whose path intersects the region of interest, wherein low frequency components of the image are determined from the incident radiation at the first radiation level whose path intersects the region of interest and the incident radiation at the second radiation level that does not intersect the region of interest.

9. The method of claim 6, further comprising arranging a pre-patient filter between the x-ray transmitter and the adaptive collimator device, wherein the pre-patient filter comprises a mechanical filter that is controllable by the controller device to reduce the radiation level and focus a radiation beam from the x-ray transmitter on the region of interest.

10. The method of claim 6, further comprising arranging a pre-patient filter between the x-ray transmitter and the adaptive collimator device, wherein the pre-patient filter comprises a static filter that focuses radiation from the x-ray transmitter on the region of interest.

11. The method of claim 6, wherein a size of the opening and/or a positioning of the opening is adjusted during each rotation of the x-ray transmitter around a subject.

12. The method of claim 6, further comprising selecting the region of interest before emitting a radiation beam.