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(54) **METHOD AND SYSTEM FOR 4D
RADIOLOGICAL INTERVENTION
GUIDANCE (4D-CATH)**

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(57) **ABSTRACT**

An imaging method for radiologically guiding an instrument during medical interventions on an object comprising: providing a first image of said object followed by providing updated images on-the-fly during the intervention to an operator by measuring an undersampled set of projections of said object and reconstructing said updated image based on changes between said first image or an update of said first image and said undersampled set of projections. The invention further relates to particular uses of the method and a system for radiologically guiding medical interventions on an object, comprising
means to provide a first image of the object;
an imaging apparatus measuring undersampled sets of projections;
processing means in communication with the imaging apparatus for providing updated images on-the-fly during the intervention by reconstructing said updated image based on changes between said first image or an update of said first image and said undersampled set of projections.

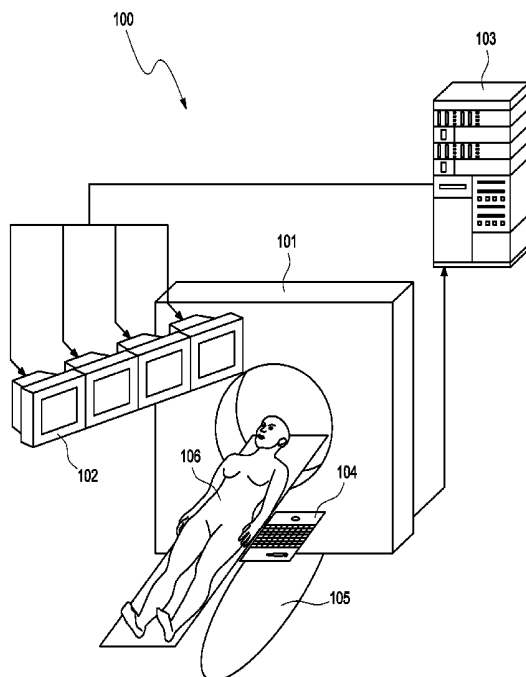
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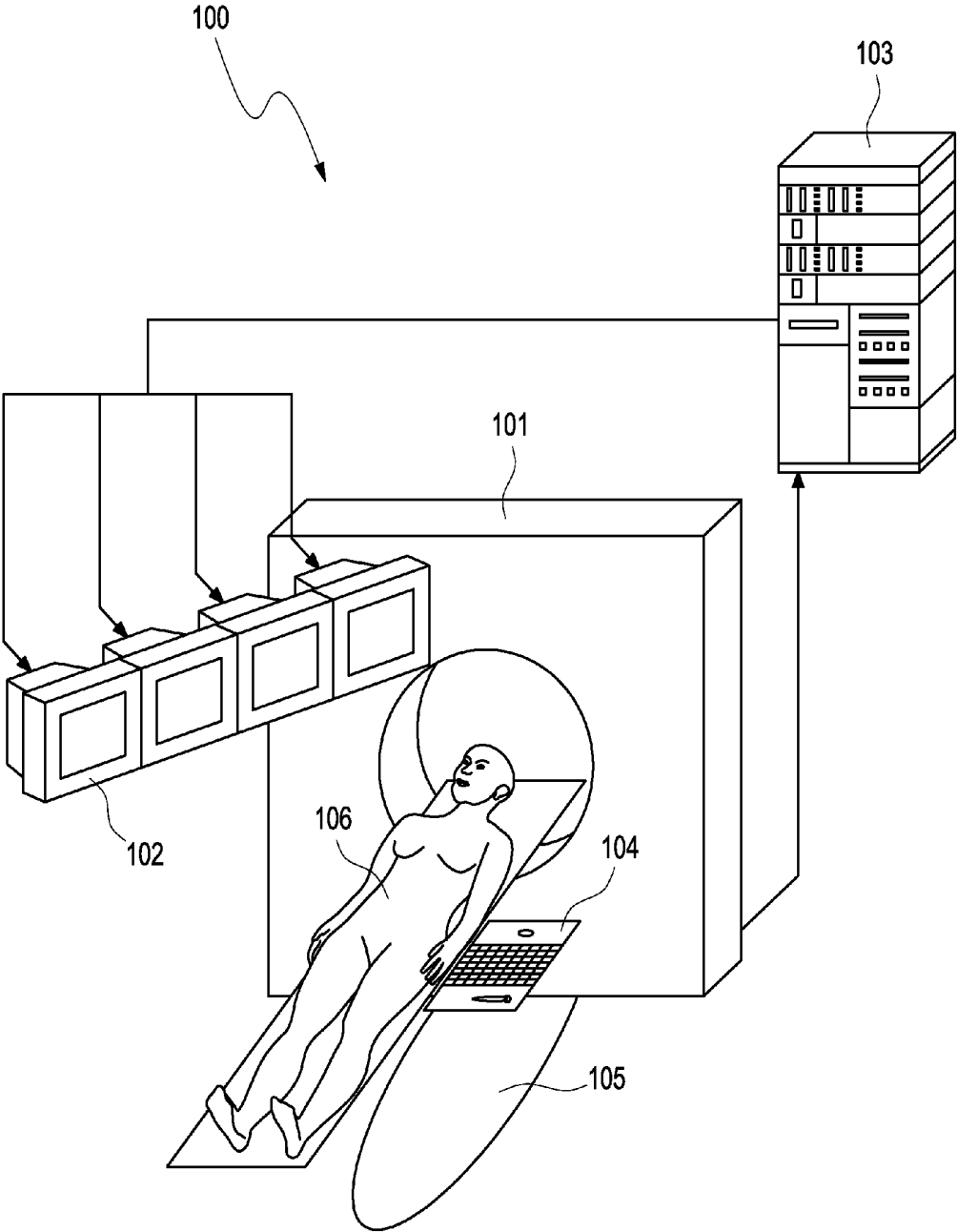


Fig. 1

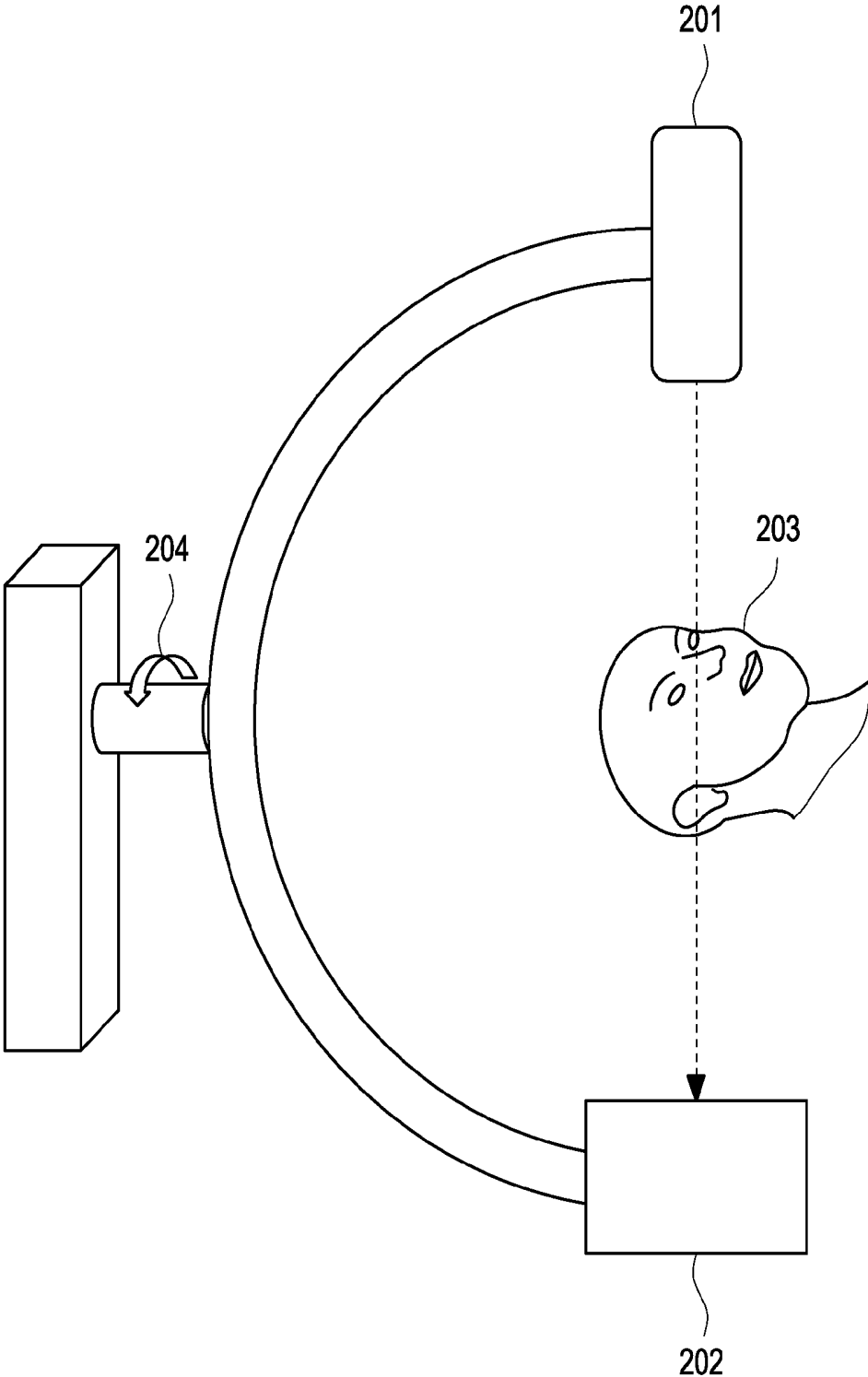


Fig. 2

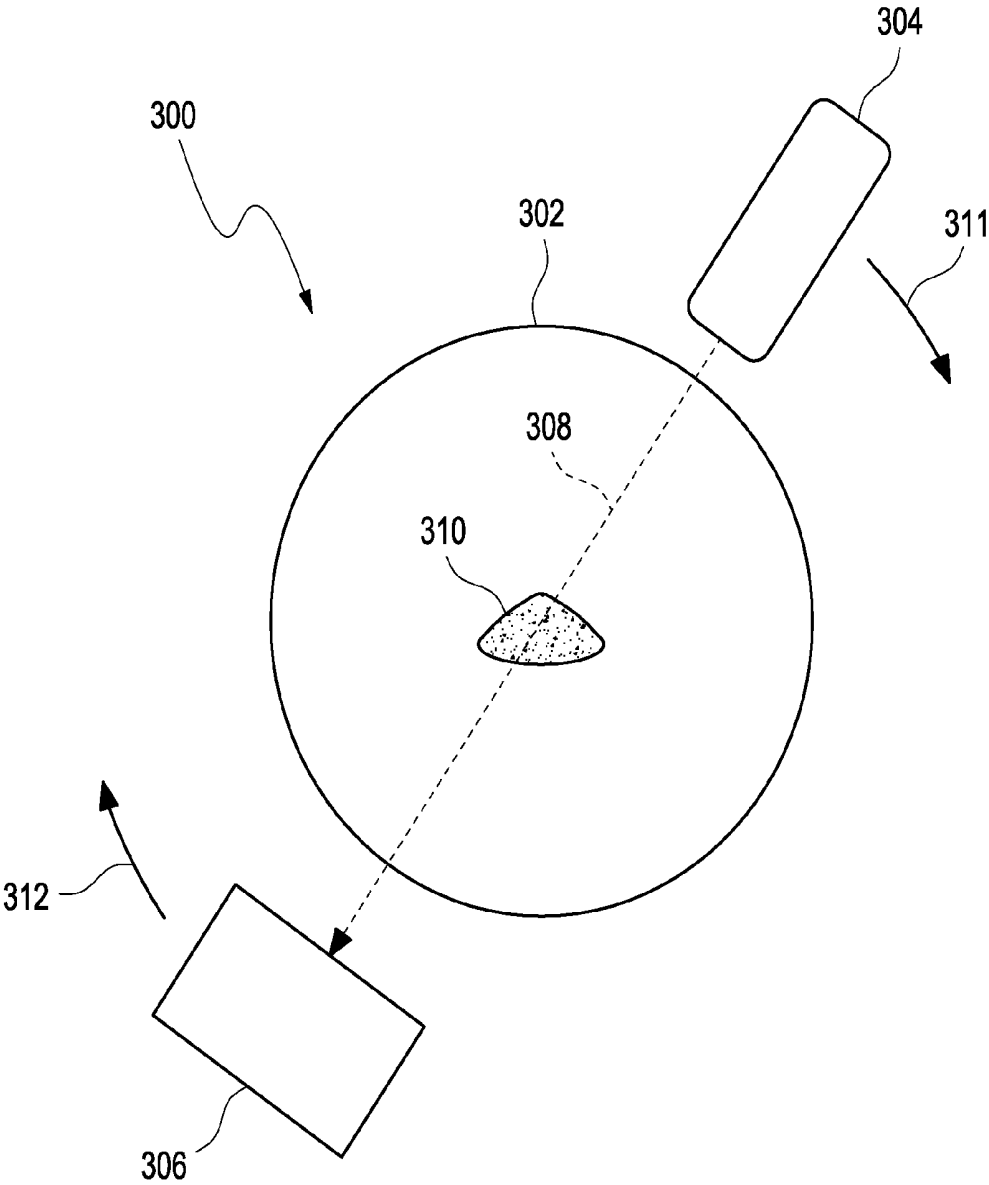


Fig. 3

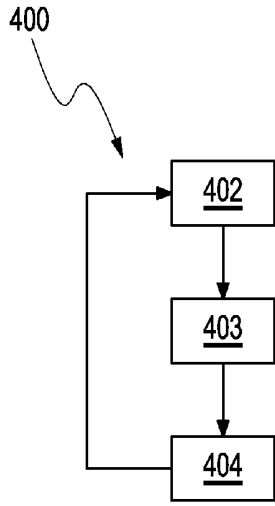


Fig. 4.1

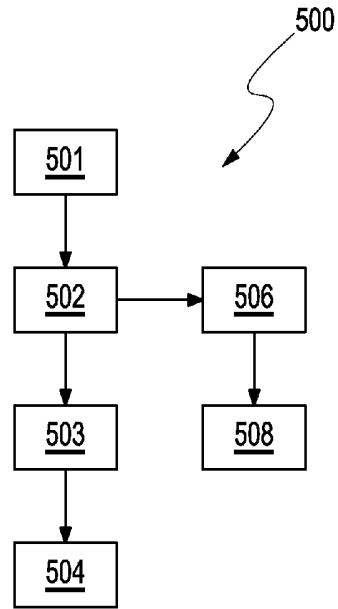


Fig. 4.2

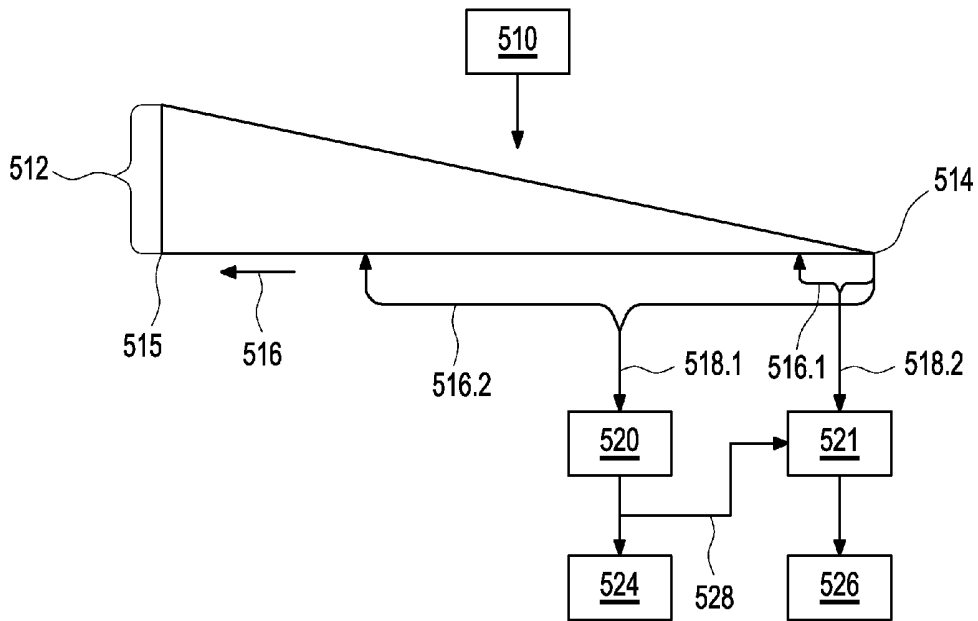


Fig. 5

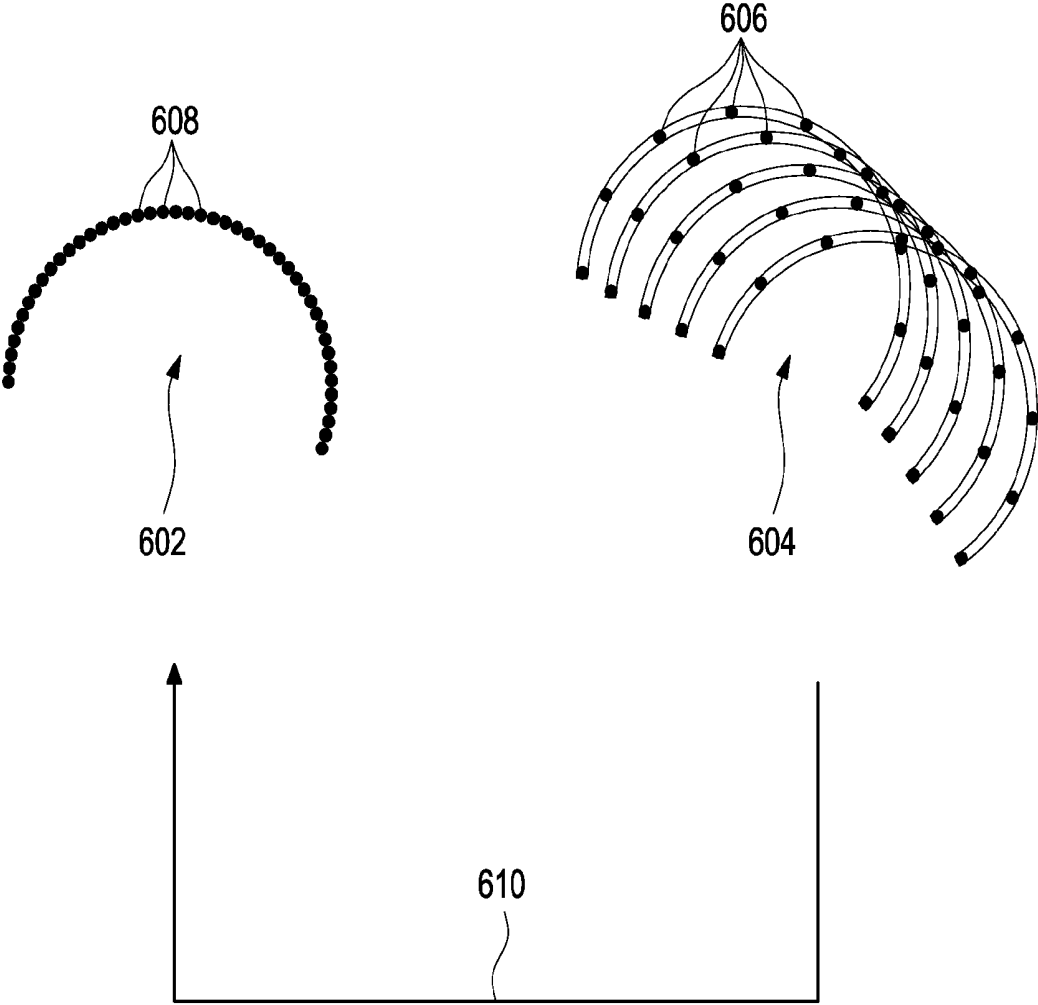


Fig. 6

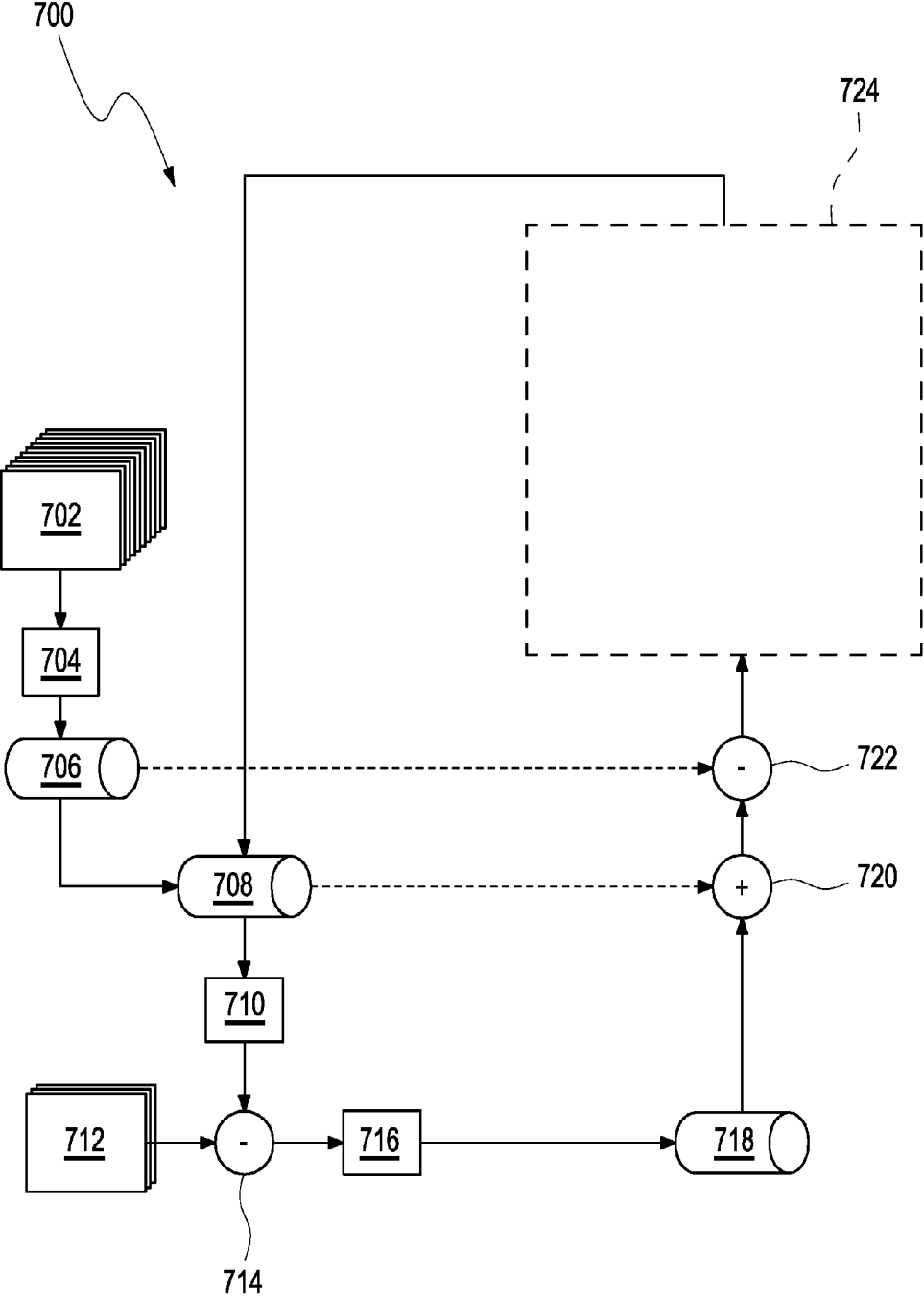


Fig. 7

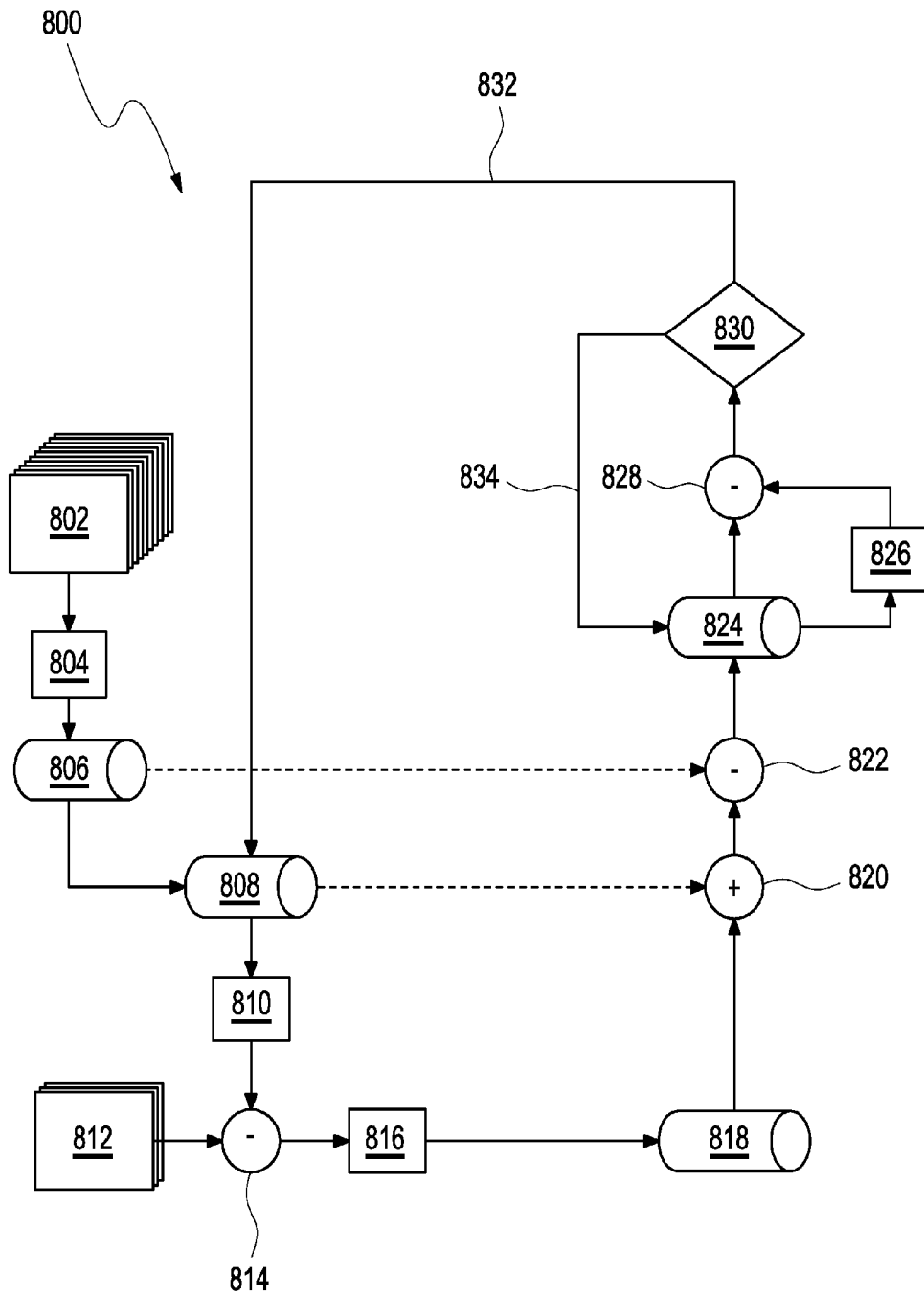


Fig. 8

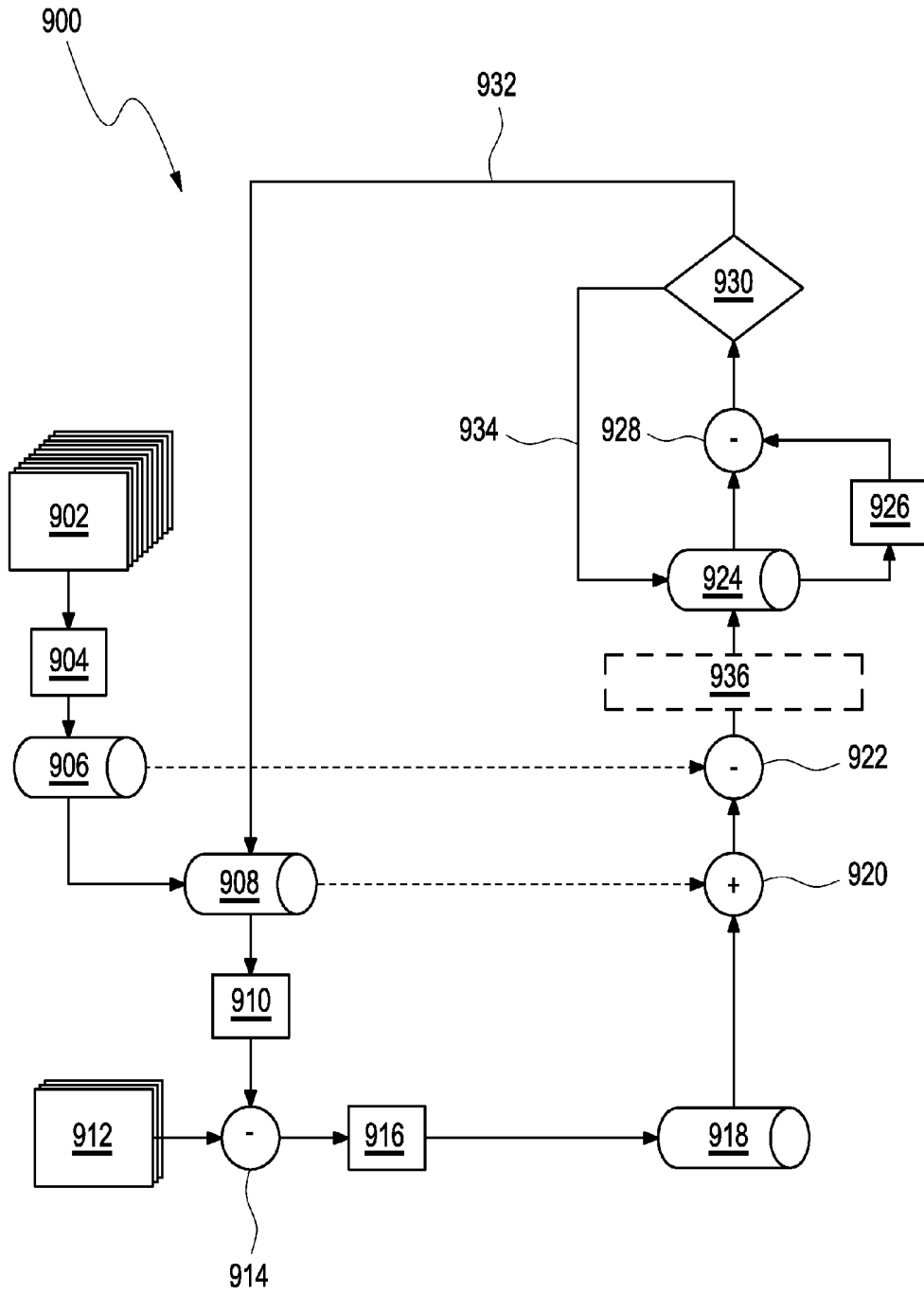


Fig. 9

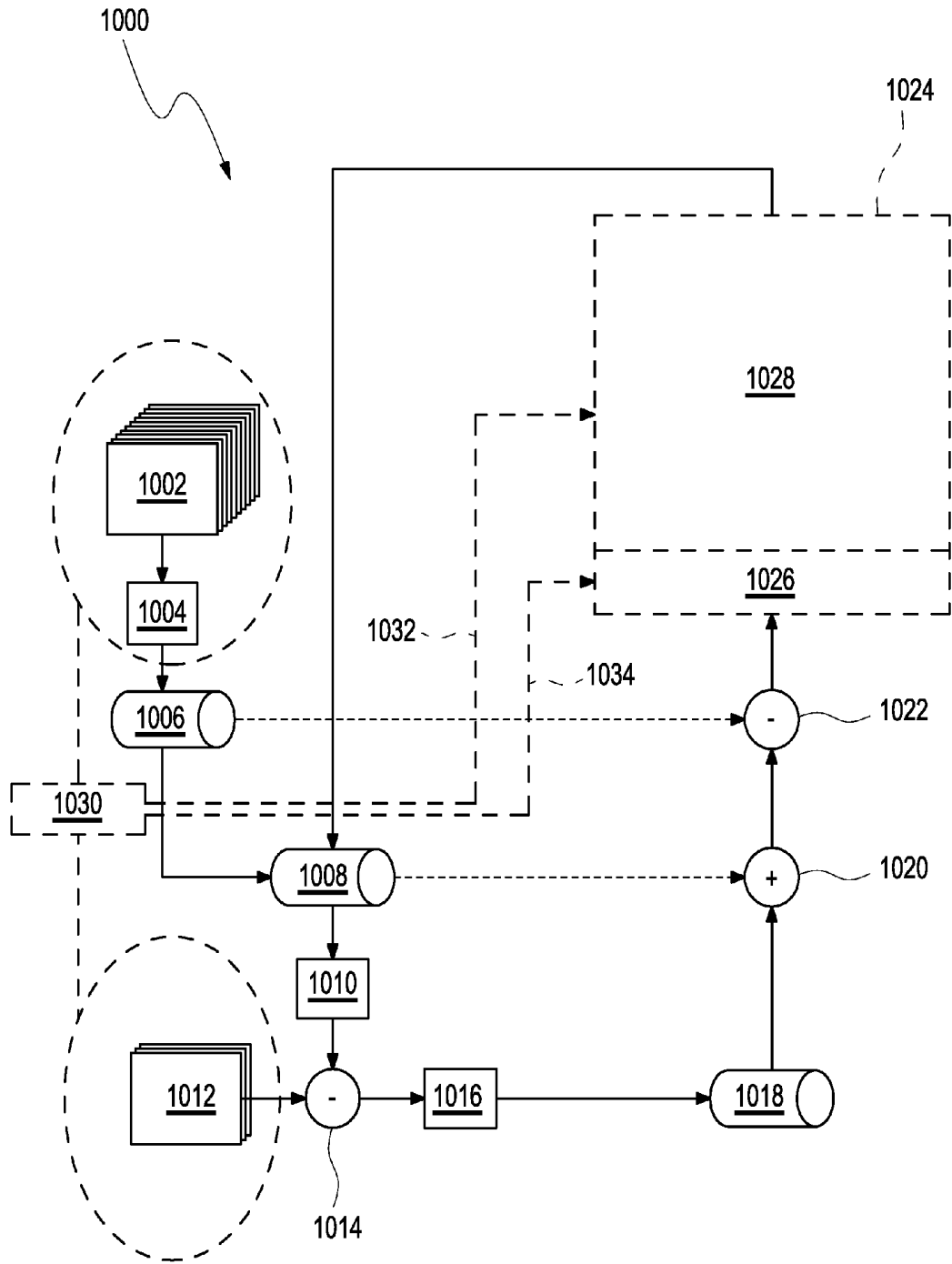


Fig. 10.1

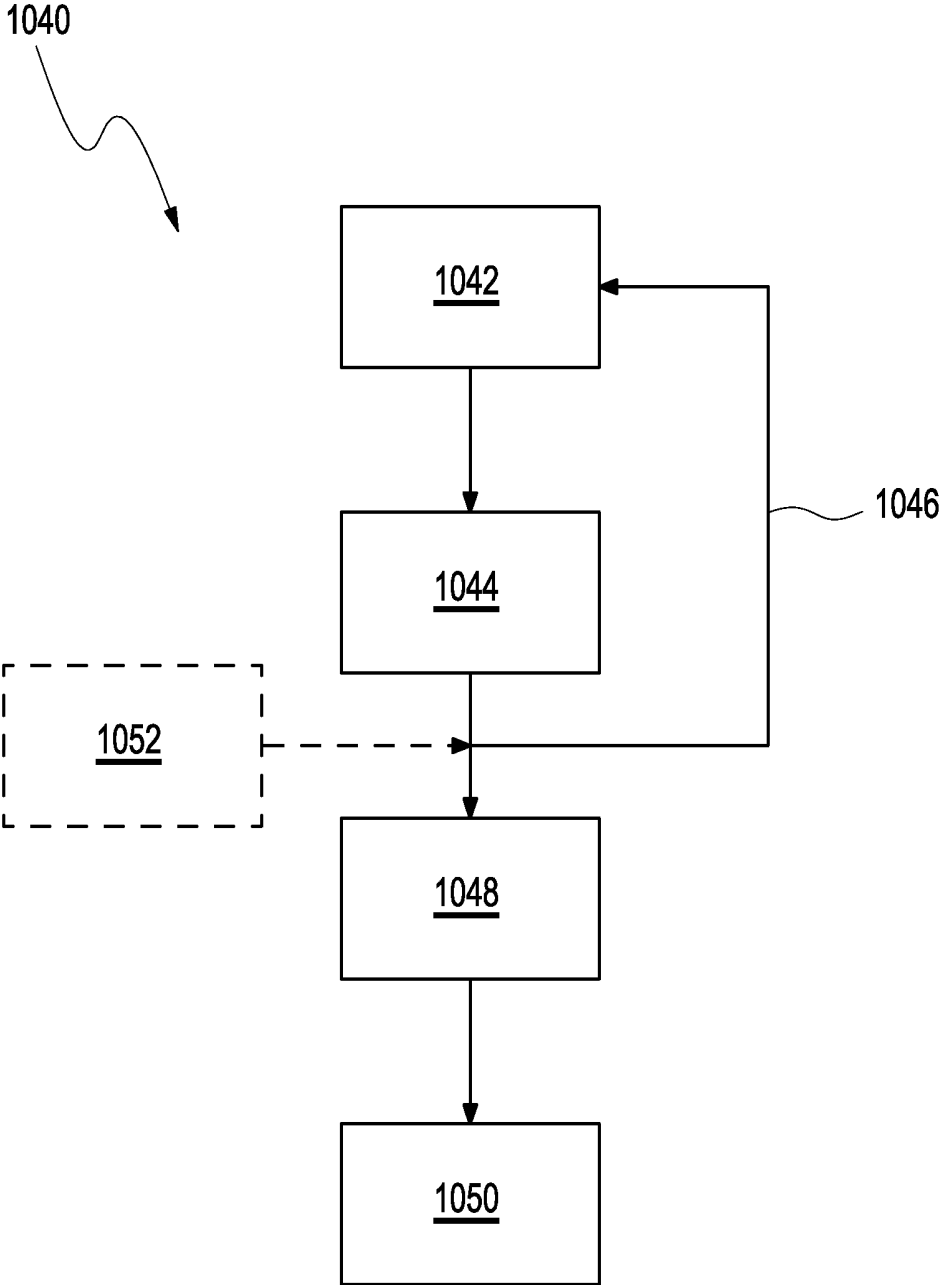


Fig. 10.2

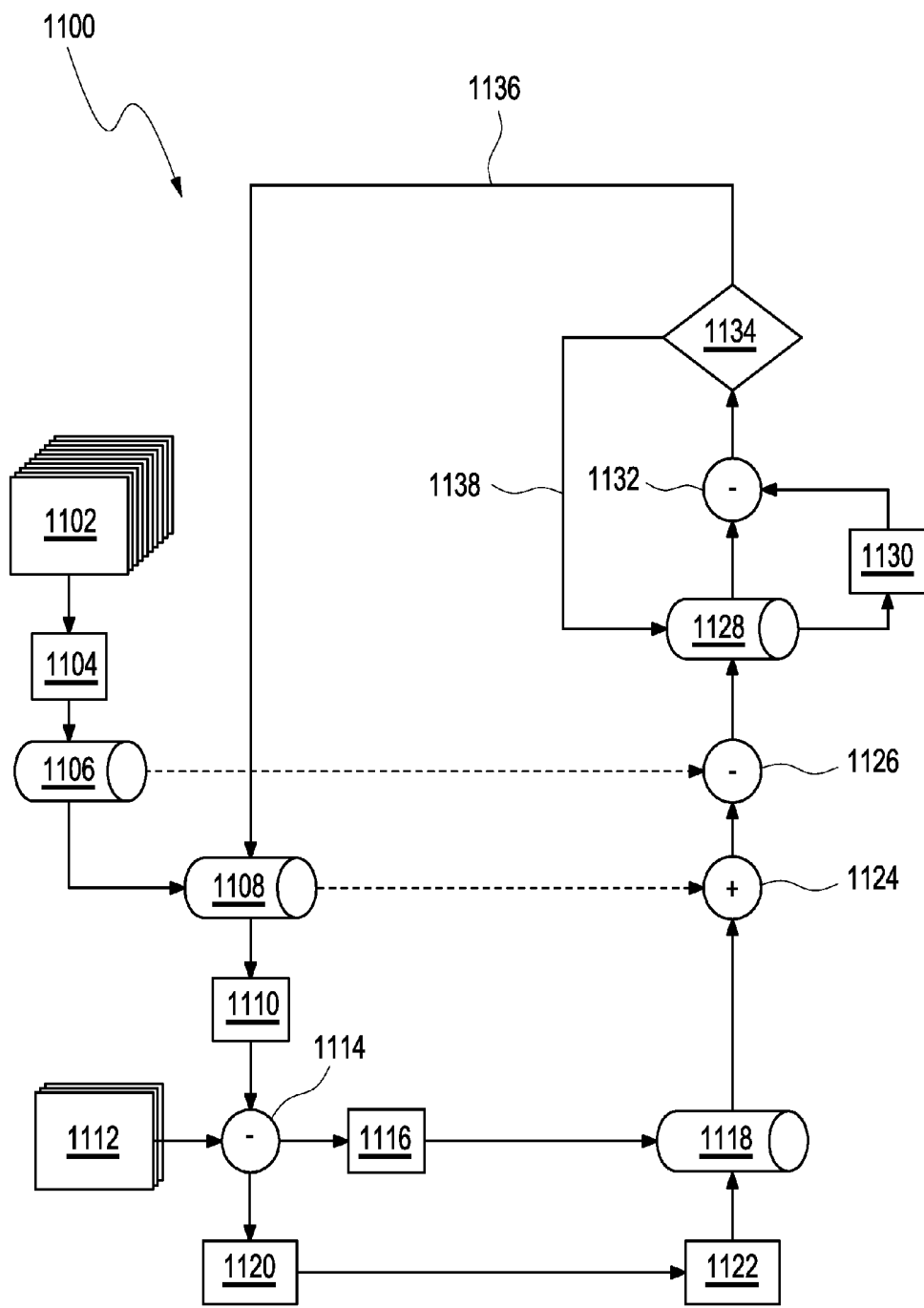


Fig. 11

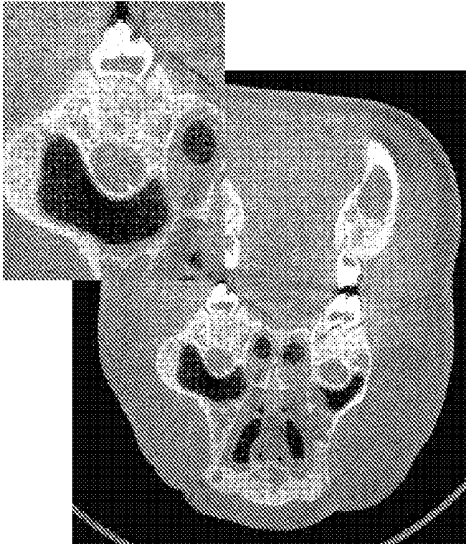


Fig. 12.1

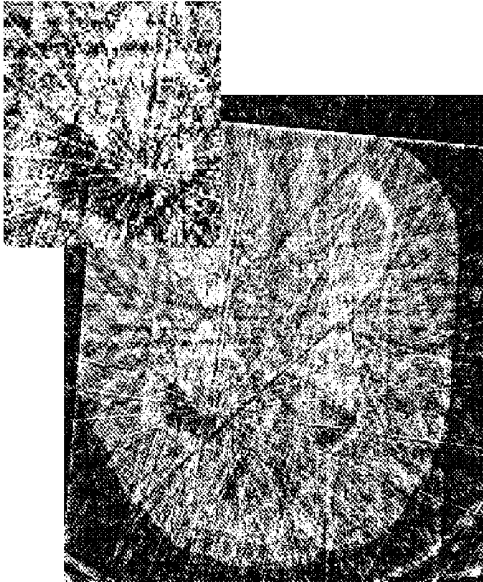


Fig. 12.2

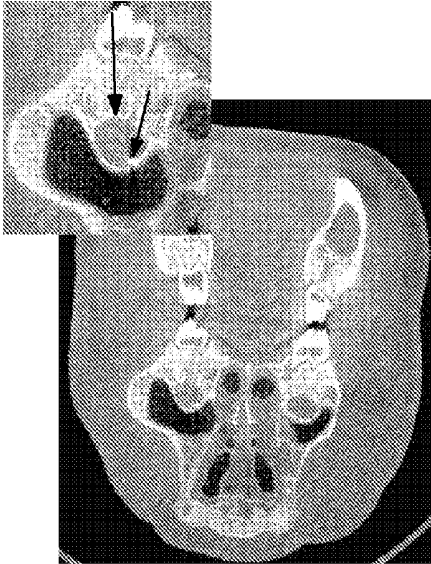


Fig. 12.3

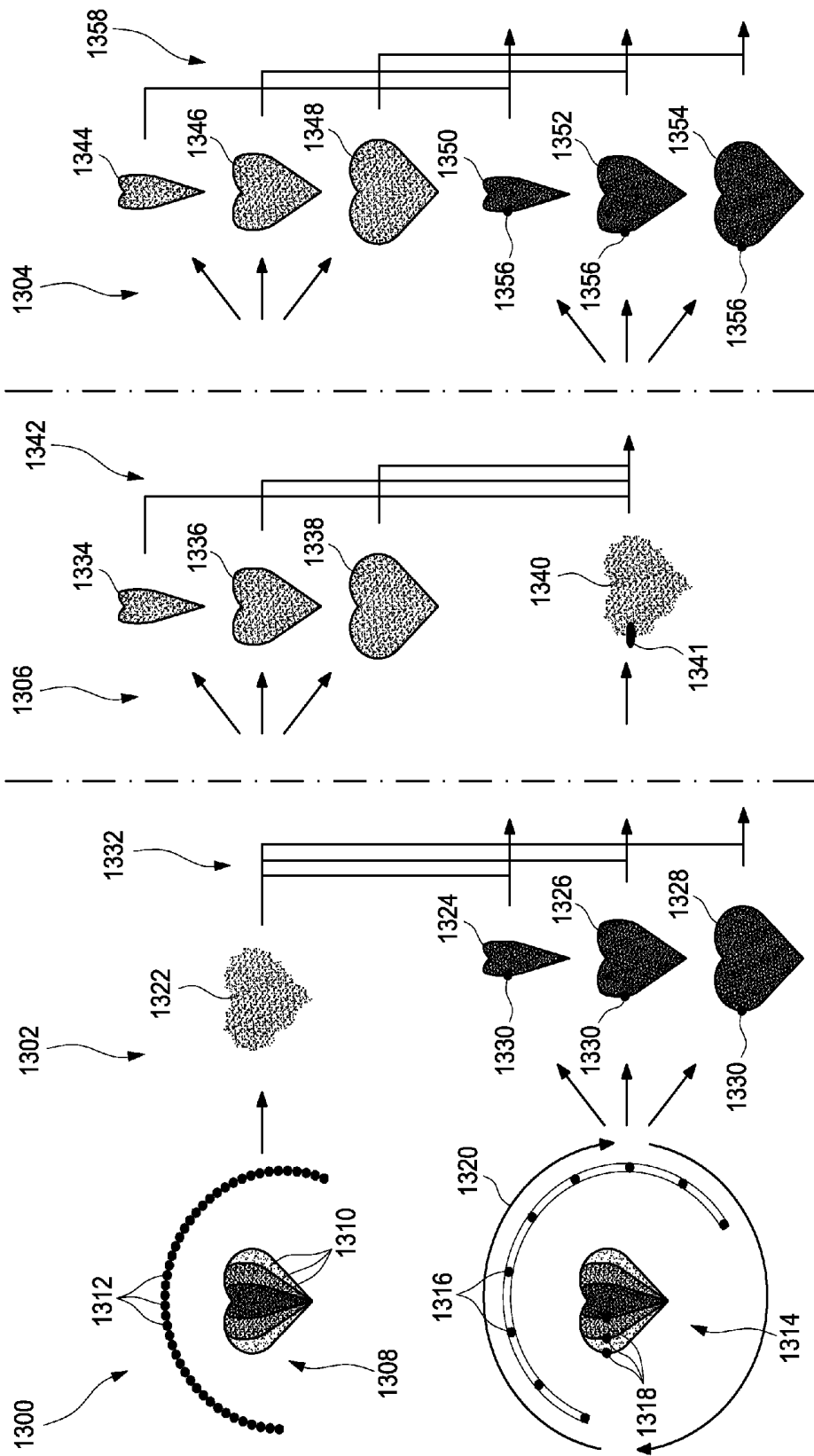


Fig. 13

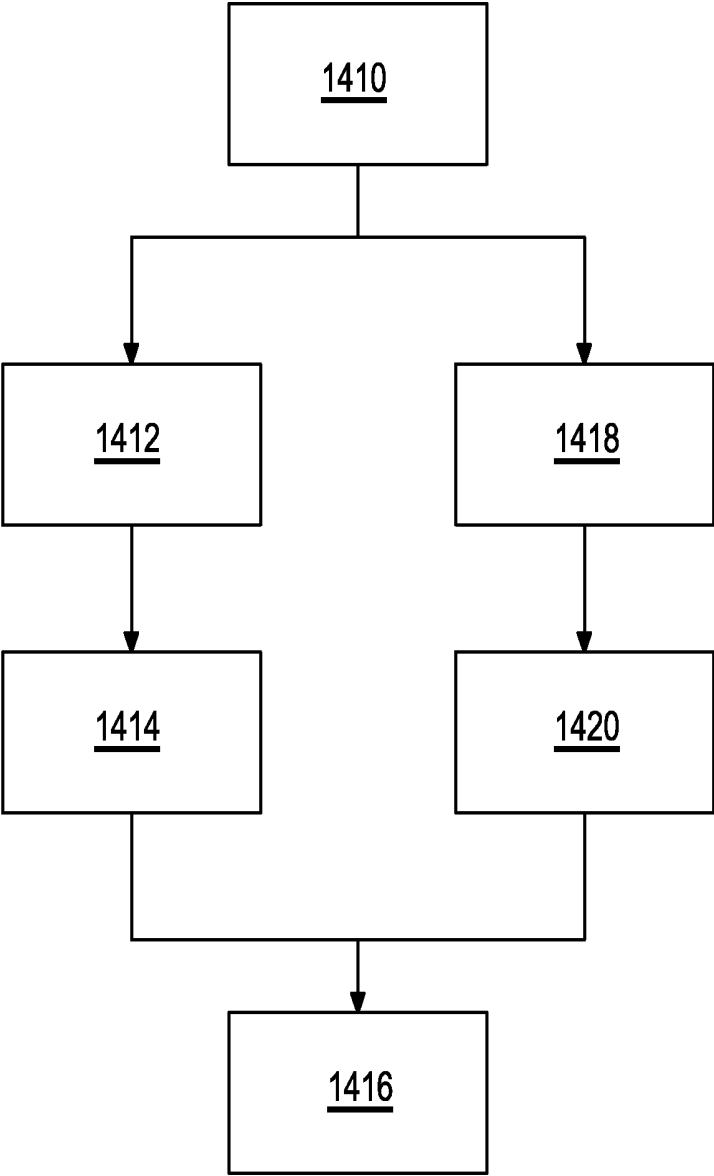


Fig. 14

METHOD AND SYSTEM FOR 4D RADIOLOGICAL INTERVENTION GUIDANCE (4D-CATH)

FIELD OF INVENTION

[0001] The present invention relates to an imaging method for radiologically guiding an instrument during medical interventions on an object. The invention further refers to a system for radiologically guiding medical interventions on an object.

[0002] Radiologically guided interventions are currently limited by known imaging methods, because known imaging methods do not allow 4D imaging (3 spatial dimensions plus time), while radiological interventions are a 4D process. Current means of intervention guidance are either continuous projective imaging (X-ray fluoroscopy) or manipulate-and-shoot computed tomography (CT) imaging. Both methods leave the interventionist with a high degree of uncertainty regarding the position of his instruments and the current surroundings.

[0003] Recently, there have been approaches for using magnetic resonance (MR) to guide interventions. C. O. Shirra et al. (Magnetic Resonance in Medicine 62:341-347, 2009) describe a framework for visualization of active catheters in 3D. In this framework compressing sensing is employed to gain high undersampling factors. Here constraints are introduced taking into account prior knowledge of catheter tube geometry and catheter motion over time to improve and accelerate image reconstruction. Furthermore, given the incremental motion of the catheter, the known position of the device of the preceding timeframe is used to penalize the reconstruction problem. In particular, the method of Schirra et al. includes the acquisition of data with randomly undersampled phase encodes; image reconstruction through constrained compressed sensing reconstruction and curve fitting the catheter outline, which can be displayed on a high resolution roadmap. Nevertheless, MR imaging still requires special arrangements to detect and visualize the instruments used for interventions. In this respect CT is due to the nature of X-Rays easier to handle, but present CT techniques such as CT-fluoroscopy require excessive radiation doses, preventing them from being routinely used.

[0004] For the reconstruction of X-ray CT images different reconstruction frameworks with sparsely sampled view angles have been proposed in order to reduce the radiation dose for the patient. In G.-H. Chen et al. (Medical Physics 35 (2), February 2008) the PICCS algorithm is presented, which reconstructs target image sequences of one measured dataset and a prior image reconstructed from the same dataset. In the reconstructed prior image the dynamic information is lost, but static structure in the image are well reconstructed. The prior image is therefore used to subtract the static structure from the target image for each time frame. A further sparsifying transform (discrete gradient) is used to further sparsify the difference. For the reconstruction of dynamic data, the data set is further gated using the simultaneously recorded electrocardiogram.

[0005] E. Y. Sidky et al. (Physics in Medicine and Biology 53, 4777, 2008) and J. Bain et al. (Physics in Medicine and Biology 55, 6575, 2010), disclose reconstruction algorithms based on constraint total-variation (TV) minimization to reconstruct data with sparsely sampled view angles. Here, the optimization selects the image with the minimum TV amongst those that satisfy a given constraint.

[0006] In G.-H. Chen et al. (IEEE Transaction on Medical Imaging PP, 1-1, October 2011), a data acquisition scheme and an imaging reconstruction method for time-resolved cardiac imaging particularly relevant to image guidance for accurate procedural planning and cardiac functional evaluations is described. The data acquired during a single gantry rotation of a C-arm system is reconstructed to a prior image using the classical FDK algorithm (filtered back projection algorithm according to Feldkamp, Davis and Kress). Then the data is gated into the different cardiac phases resulting in undersampled sets for each phase, that are reconstructed through the prior image constraint compressed sensing (PICCS) algorithm utilizing the reconstructed prior image.

[0007] H. Langet et al. (MICCIA 2011, Part I, LNCS 6891, p 97-104, 2011) discloses a method for 3D reconstruction of rotational angiography based on an iterative filtered back-projection approach that includes a sparsity constraint called soft background subtraction. This approach is particularly useful when injecting a contrast medium particularly leading to late vessel opacification. Z. Qi et al. (Physics in Medicine 56, 6709, 2011) discloses a 4D cone beam CT method which enables the detection of e.g. respiratory motion. The acquired data is binned into the different respiratory phases and reconstructed using PICCS. From the reconstructed 4D cone beam computed tomography images, the motion trajectory for an object is extracted using deformable registration methods.

[0008] However, there is still a need for real time CT imaging for guiding interventions, which reduces the exposure of the patient to ionizing radiation to a minimum.

OBJECT OF THE INVENTION

[0009] The present invention aims to provide a method for true 4D imaging which overcomes above mentioned limitations to improve radiologically guided interventions. It is a further object of the present invention to provide a method for fully CT guided medical intervention providing three-dimensional information in real time during the intervention. The present invention particularly aims at providing 4D CT imaging that avoids excessive radiation dose for the patient and can be routinely used in medical applications.

[0010] According to the invention, an imaging method for radiologically guiding an instrument during medical interventions on an object is proposed, which comprises the following steps:

[0011] providing a first image of said object followed by
[0012] providing updated images on-the-fly during the intervention to an operator by measuring an undersampled set of projections of said object and reconstructing said updated image based on changes between said first image or an update of said first image and said undersampled set of projections.

[0013] Additionally, according to the invention, a system for radiologically guiding medical interventions on an object, according to the method previously explained, is proposed. Such a system comprises:

[0014] means to provide a first image of the object;

[0015] an imaging apparatus measuring undersampled sets of projections;

[0016] processing means in communication with the imaging apparatus for providing updated images on-the-fly during the intervention by reconstructing said updated image based on changes between said first image or an update of said first image and said undersampled set of projections.

[0017] The proposed imaging method and imaging system for radiologically guiding the instrument during medical

interventions allow for dynamic real-time imaging during the course of the intervention. Thus, volumetric data is acquired in close, timely consecution allowing the operator to control and monitor the course of the intervention and particularly the movement of the instrument in three spatial dimensions. The proposed method and system for medical imaging aim to provide the temporal changes in the object to be imaged, such as instrument movements, by measuring a small number of projections for the undersampled sets of projections. This way, the radiation dose, the patient is exposed to, can be reduced to a minimum, while still providing sufficient information on temporal changes, such as instrument movement.

[0018] Furthermore, the proposed imaging method and imaging system allow an operator to monitor and hence to control the intervention. Such interventions can be performed on body parts, such as the cardiovascular system, tubular organ structures or on the brain, which exhibit complicated three-dimensional structures. In particular, the time component, i.e. the dynamics of the intervention, play an important role in order to ensure safe performance. This provides the interventionist with a high degree of certainty in the way the intervention is performed and thus, reduces the risk for the patient during intervention.

[0019] Thus, the proposed method facilitates fully guided interventions based on imaging, which is particularly relevant to interventions such as catheter interventions, bronchoscopy interventions, implantation of cardiac pacemakers or positioning of stents. Hence, maneuvering in complicated three-dimensional structures becomes easier and misplacement of e.g. stents or injury due to e.g. rupture of vessels can be avoided. Overall, the imaging method according to the present invention provides full control in three-spatial dimensions including temporal changes during the course of the intervention.

[0020] In contrast to known systems, where the projections that are used for guidance cannot be used for tomographic reconstruction, the invention allows to use the projection data for the reconstruction of several pieces of information relying on the same data acquisition. This significantly reduces the relative radiation costs. Additionally, through the application of tomographic imaging a lower concentration of contrast media is detectable.

[0021] Furthermore, the invention allows novel intervention concepts to become feasible. For instance, biopsies can be combined with endoscopic or intravascular accesses. The intravascular access route can be used for biopsies of extravascular structures, e.g. pancreas biopsies through the superior mesenteric artery. Extrabronchial masses can be accessed through a bronchus, while the needle towards the mass is then imaged using the said tomographic imaging. In particular at present, no tomographic imaging is available, which is why the biopsy of extraluminal lesions using endoscopes is limited to the optically, visible wall lesions.

[0022] In the sense of the present invention, imaging refers to tomographic imaging providing a volumetric image of an object as reconstructed from one-dimensional or two-dimensional projections. A projection hereby represents a two-dimensional (once integrated) or a one-dimensional (twice integrated) image of a three-dimensional distribution at specific projection solid angles of the source-detector assembly with respect to the object to be imaged.

[0023] In particular, medical imaging is a technique and process to create images of the human body for clinical purposes. The object to be imaged can either comprise the full

body of a patient or parts and functions thereof. Typical imaging techniques used in this area are tomographic imaging techniques, such as magnetic resonance imaging (MRI) or computer tomography (CT). In the case of MRI non-ionizing radio frequency (RF) signals are used to acquire images. CT, on the other hand, uses X-rays (a type of ionizing radiation) in order to image objects. Owing to the dose X-rays carry, only a restricted number of exposures can be performed and the number of projections to be measured is to be kept at a minimum. However, instrument detection via X-rays is due to the nature of radiation simpler than instrument detection through MR-signals. In a preferred embodiment of the invention, said imaging method can be based on ionizing radiation particularly X-rays, wherein the undersampled set of projections is measured with a minimized radiation dose.

[0024] The method according to the present invention provides updated images "on-the-fly", meaning measurements, calculations and modifications take place without significant time delay during the intervention. Hence, the method provides a way of real-time imaging during medical interventions. Furthermore, the real time images are provided to an operator, who can be the interventionist or a medical robot, to further support the course of the intervention. Lastly, an updated image represents an image that includes the latest temporal changes recorded during the intervention.

[0025] In the sense of the present invention, an "undersampled set of projections" represents a sparsely sampled set of projections, meaning the number of measured projections violate the Shannon-Nyquist sampling theorem. In view of the present invention, the Shannon-Nyquist theorem is in the image domain given by the highest represented frequency f . The sampling rate must be larger than $2f$ to fulfil the Shannon-Nyquist theorem. A pixel corresponds to a data point in a matrix representing the image. In contrast to an undersampled set a fully sampled set fulfils the Shannon-Nyquist sampling theorem and includes an appropriate number of projections essentially uniformly distributed between 0° and $(180^\circ + \text{cone angle})$. Here essentially uniformly encompasses deviations of up to 10%. In practical interventional radiology, the under-sampling factor can lie in the order of 10-30 resulting in 8-35 frames per reconstruction.

[0026] In one aspect of the present invention, said first image, said update of said first image and said updated images comprise volumetric images of the object to be imaged. Here, the first image, which includes the static structure of the object may be reconstructed from at least one undersampled set of projections, at least one fully sampled set of projections or a combination hereof. Particularly, said first image can comprise a high-resolution volumetric image of the object, which can be acquired prior to the intervention. Such a high resolution volumetric image can be produced by measuring a fully sampled set of projections using CT imaging or any other medical imaging modalities. Another option comprises to provide a high resolution volumetric image from a database, where for example previously recorded images of the object to be imaged are stored. This way a reconstruction for Prior Image Dynamic Computed Tomography (PRIDICT) can be realized allowing for real-time interventional guidance.

[0027] In a further aspect of the invention, a first image reconstructed from an undersampled set of projections can be improved during the course of the invention by incorporating any projections measured to reconstruct the updated image into the first image leading to an updated first image. The

update of the first image thus encompasses the first image and at least parts of one or more updated images reconstructed in previous runs during the intervention. Previous runs in this context comprise any reconstructed image or equivalently any projections that have been acquired at some time before the current updated image. Thus, the iterative method provides updated images in real-time, where a reconstruction performed at a point T_2 in time may include an update of the first image, which again, comprises any updated images computed at earlier points in time $T_1 < T_2$.

[0028] In another aspect of the present invention, the update of the first image is a sliding prior defragmenting updated images of previous runs. Thus, the update of the first image at a point T_2 can incorporate updated images produced at any point $T_1 < T_2$, wherein the update of the first image is successively updated for every reconstruction performed. In this embodiment, the projections of the update scans are preferably acquired at projection angles that are different from earlier projection angles such that after several rotations a new-fully sampled dataset is produced, which can be used as update of the first image. This is particularly advantageous when movement of the object occurs during the intervention. Such movement may e.g. include displacement of the patient relatively to the imaging system.

[0029] In another embodiment of the present invention, not only changes, which progress fast in time, such as catheter movement, but also slower changes, such as bleedings, may be visualized by storing the undersampled sets of projections measured on-the-fly during the intervention for a delayed reconstruction of soft tissue contrast.

[0030] In another aspect of the invention, the method for reconstructing the updated image based on changes between the first image or the update of the first image and the undersampled set of projections is proposed, which utilizes a compressed sensing framework. This framework is generally known for application in CT (see e.g. E. Y. Sidky et al., *Imaging Reconstruction in Circular Cone-Beam Computer Tomography for Constrained, Total-Variation Minimization*, *Phys. Med. Biol.*, 2008, 53, 4777-4807). In general, compressed sensing allows for image reconstruction from randomly undersampled data violating the Shannon-Nyquist sampling theorem. A necessary condition for a successful reconstruction thereby is the sparsity of the given data in any transform domain, such as the image domain. For the highest represented frequency f in the image domain and the sampling rate corresponding to the number of measured projections D , image recovery is possible as long as $D < 2f$. Assuming the projections are measured at random projection angles, the image can be recovered from D measurements with high probability by solving the L_1 optimization problem, wherein L_1 denotes the norm defined by the sum of the absolute values in each pixel.

[0031] In one realization the reconstruction algorithm according to the present invention is an iterative reconstruction algorithm. In a one step of this algorithm the amount of changes between different images taken at different points in time during the intervention can be determined by forward projecting the first image or the update of the first image according to the projections comprised in the undersampled set of projections. Thus, a set of equivalent projections based on the first image or on the updated of the first image can be provided in order to perform a subtraction operation between equivalent projections of the undersampled set of projections and the projections resulting from the first image or its update.

Here, equivalent projections refer to forward projected images having the same projection angles.

[0032] From the difference projections, a difference volumetric image may be reconstructed. Through the subtraction operation, static parts of the object to be imaged, which are included in the first image or the updated first image, are effectively subtracted from the measurement characterizing the updated image. Therefore, the updated image basically represents the temporal change in the object to be imaged. During, for instance catheter interventions, this temporal change is given by the movement of the instrument.

[0033] In another implementation of the invention, the reconstruction includes an iterative minimization of a number of significant pixels, optionally including further sparsifying functions, which might be applied after reconstruction of the differences. The number of significant pixels thereby includes all pixels signifying the temporal changes, and thus, representing the dynamic rather than the static structure of the image. Further sparsifying functions may include gradient operations, wavelet transformations, curvelet transformations, contourlet transformations or a combination thereof. Such transformations aid to reduce the number of significant pixels by further sparsifying the image to be reconstructed.

[0034] In a further implementation of the reconstruction algorithm, the algorithm itself or the scan parameters can be influenced by the amount of changes between successive measurements of projections. A combination of different sparsifying transforms and/or one tunable sparsifying transform can be used in different configurations during one reconstruction to correctly reconstruct different structures, e.g. point-like or curve-like structures. Alternatively or additionally, the weight of different sparsifying transforms can be varied e.g. according to the sparseness of the transformed image. In particular, the influence by the amount of change on the algorithm can include a further sparsification of the image, if the changes are less significant with time.

[0035] Alternatively, the sparsification may solely be based on the difference excluding further sparsification, if there are significant changes in time. Significant temporal changes are, for example instrument movements, in contrast to less significant temporal changes, which may be related to vessel bleedings or the like. Here, significant changes can signified by the rate of change, which may be faster than one changing feature corresponding to one or more changing pixels in the image per minute, preferably per 30 seconds. In contrast, a less significant change may constitute less than one changing feature corresponding to one or more changing pixels in the image per minute, preferably per 1.5, particularly preferably per 2 minutes. Furthermore, the scan parameters such as the number of projections in the undersampled set of projections and/or the dose rate per projection can be adapted according to the amount of changes between measurements and/or according to an input provided by the operator.

[0036] In a further implementation of the present invention, the at least one further sparsifying function included in the reconstruction depends on the amount of data that has been acquired before the actual present. In this context, the actual present signifies a range in time, wherein the longer ago from the present the higher the amount of low-dose tomographic data that has been acquired during the intervention. Thus at a point, where the amount of data for tomographic reconstruction only comprises a few projections, the amount of data is small and updated images can be reconstructed excluding sparsifying functions, preferably by minimization directly in

the image domain through PRIDICT (Prior Image Dynamic Computed Tomography). If the amount of data for tomographic reconstruction comprises more projections, for example data stored from previous runs, images can be reconstructed with sparsifying function. In particular, updated images can be reconstructed with no further sparsifying function. Additionally or alternatively, the reconstruction of data stored from previous runs can include at least one sparsifying function.

[0037] In yet a further implementation of the algorithm, the reconstruction of the updated image includes motion compensation. Motion compensation describes a method, which tracks motion in successive images and formulates the motion in terms of a transformation of one image to a successive image or vice versa. Such motion compensation is usually based on comparing corresponding features in successive images and identifying motion vectors, which may be used to build up a motion vector field. In particular, through motion compensation periodic and/or non-periodic motion of the object or a structure within the object to be imaged can be compensated for in the reconstruction of the updated images. Specifically, periodic motion compensation can be performed through gating measured projections into different phases of the periodic motion and utilizing the gated projections for reconstruction. For gating the cycle of the periodic motion is typically fragmented in to several intervals signifying the phases of motion and the projections are distributed into the corresponding intervals during measurement. In interventions periodic motions can result from the respiratory and/or the cardiac motion. During the course of the intervention, respiratory and/or cardiac motion are recorded providing a type of reference signal for the gating. Measured projections are then distributed into the different phases according to that reference signal.

[0038] In another implementation periodic motion compensation can be performed through a transformation mapping of images into one phase of the periodic motion. This can be realized by applying deformations via motion-vector fields. Examples for such algorithms are the McKinnon-Bates algorithm or the Phase-Related Feldkamp algorithm. In comparison to gating, transformation mapping enables to use all the measured data for reconstruction, which further minimizes the radiation dose during intervention and still provides high image quality at all time points of a motion cycle. In another implementation the motion compensation will be performed by recording a fully sampled 4D dataset of the imaged object, e.g. the heart or the lung. From this data set motion vectors are calculated that can then be used to support the motion compensated reconstruction.

[0039] The updated images reconstructed iteratively during the course of the intervention can be displayed in real-time on a screen allowing for different representation modes which are chosen automatically or by the operator. The image diagnostically relevant for the physician is the sum of the first image including static structures and the temporal update giving the updated image. Therefore, the updated imaging including static and dynamic structure can be provided to the operator, preferably through a screen. The representation of such images may for instance be preset including either a full 3D representation of the object to be imaged and further parameters, which might be of interest during intervention guidance. Here, the 3D representation may comprise some solid or boundary model of the object to be imaged and in particular the relevant structures or functions therein. The

representation mode can be selected from volume rendering, multiplanar reformations and all other means of medical image presentation. In a further version, image analyze tools are used to trace the intervention instruments in the data set and provide the interventionist with region of interest representations using e.g. curved multiplanar reformation, segmentation or similar tools.

[0040] Furthermore, with the updated image the operator can be provided with an indicator of the accuracy of the displayed image. This indicator of accuracy can depend, for instance, on the reconstruction parameters, such as matrix size or time resolution, the scan parameters, such as number of projections in the undersampled set, or the like. One indicator can be the total variation or other mean values of the displayed image to assess the difference between actual projections and forward projections through later reconstructed volumes.

[0041] According to the invention a computer program for performing above described method is proposed, when executing the computer program on a computer, particularly a high performance computing device (HPC). The computer program is preferably stored on a machine readable storage medium or on a removable CD-Rom, Flash memory, DVD or USB-stick. Additionally or alternatively, the computer program is provided on a server to be downloaded via, for example, a data network, such as the internet or another transfer system, such as a phone line or any wireless connection.

[0042] Further according to the invention, above described method is used during interventions on the cardiovascular system, during catheter intervention and/or for use in the implementation of cardiac pace makers. Further according to the invention, above described method is used during interventions on tubular organ structures, preferably lungs or kidneys, in positioning of stents in vessels or bronchi or used during bronchoscopy interventions. Further uses comprise interventions on the brain or angiography.

[0043] Additionally according to the invention, a system for radiologically guiding medical interventions on an object, according to above described method, is proposed. Such a system comprises:

[0044] means for providing a first image of the object;

[0045] an imaging apparatus measuring undersampled sets of projections;

[0046] processing means in communication with the imaging apparatus for providing updated images on-the-fly during the intervention by reconstructing an updated image based on changes between said first image or an update of said, first image and said undersampled set of projections.

[0047] Preferably, the system encompasses the necessary elements for providing the updated image to an operator, such as screens. Preferably, the system comprises the components described in the context of the method according to the invention in order to be capable of performing the same. A system of this type is particularly advantageous, since it allows monitoring and controlling movements of an instrument during medical interventions on an object in real-time.

[0048] In a preferred embodiment of the imaging system, the imaging apparatus is a tomographic system, such as a magnetic resonance imaging (MRI) scanner or a computed tomography (CT) scanner, wherein the CT scanner comprises at least one X-ray source and at least one detector. Furthermore, the X-ray sources may differ in terms of the X-ray spectra and the detectors can provide means of energy differ-

entiation. This way potential material differentiation is being used to influence the reconstruction algorithm and/or the dual-energy information is being used to extract further information on changes related to instrument movements.

[0049] In a further embodiment of the imaging apparatus, imaging parameters of the apparatus depend on the changes between said first image or an update of said first image and said undersampled set of projections. The imaging parameters of the imaging apparatus can be influenced automatically or by an operator. For the operator to monitor and control the intervention, the system can include means to provide an input and/or an output to an operator. Preferably, the output comprises at least one updated image, at least one image produced by delayed reconstruction of soft tissue contrast and/or an indicator for an accuracy of displayed images.

BRIEF DESCRIPTION OF THE DRAWINGS

[0050] The subject-matter of the invention will be explained in more detail in the following with reference to preferred exemplary embodiments, which are illustrated in the attached drawings. The drawings show

[0051] FIG. 1, an exemplary configuration of the 4D-CATH lab is shown (here 4D-CATH stands for 4D Catheter Advancement with Tomographic Help),

[0052] FIG. 2, an illustration of a C-arm CT imaging device,

[0053] FIG. 3, an illustration of the general CT acquisition process,

[0054] FIG. 4.1, a flowchart of the general CT workflow for 4D-CATH,

[0055] FIG. 4.2, a workflow of CT image acquisition and reconstruction,

[0056] FIG. 5, an illustration of a time-dependent data acquisition,

[0057] FIG. 6, an illustration of a sliding prior,

[0058] FIG. 7, a flowchart showing the general structure of the PRIDICT reconstruction algorithm that can be used in 4D-CATH,

[0059] FIG. 8, one embodiment of the PRIDICT reconstruction algorithm as shown in FIG. 7,

[0060] FIG. 9, a further embodiment of the PRIDICT reconstruction algorithm as shown in FIG. 7,

[0061] FIG. 10.1, yet a further embodiment of the PRIDICT reconstruction algorithm as shown in FIG. 7 including a method to influence the sparsifying function,

[0062] FIG. 10.2, a flowchart of the CT image acquisition and reconstruction including influencing scan and/or reconstruction parameters,

[0063] FIG. 11, yet a further embodiment of the PRIDICT reconstruction algorithm including a further method to influence the reconstruction algorithm,

[0064] FIG. 12.1-3, exemplary reconstruction of the guide wire moved in a pig's head,

[0065] FIG. 13, an illustration of the PRIDICT reconstruction algorithm including motion compensation,

[0066] FIG. 14, a flowchart for 4D intervention guidance and 3D road-mapping.

EMBODIMENTS ACCORDING TO THE INVENTION

[0067] FIG. 1 shows an exemplary configuration of the 4D-CATH lab including a CT scanner **101** in communication with a high performance computing device (HPC) **103** further

in communication with one display or an array of displays **102** to provide the operator **105** with imaging information for guiding the intervention. The tomography system **101** is directly connected to the HPC **103** like conventional clusters, GPU-systems, GPU-clusters, cloud systems or other mainframes, where the actual reconstruction of images is performed. Thus, the HPC **103** receives projections measured by the CT scanner **101** and sends reconstructed, updated images to at least one of the displays of the array of displays **102**.

[0068] In an exemplary embodiment, the CT scanner **101** comprises a continuously rotating, gantry-based CT scanner **101** with a flat-panel detector. Such a system is for instance described in R. Gupta et al. (Flat-panel volume CT: fundamental principles, technology, and applications. Radiographics. 2008; 28(7):2009-2022). Other embodiments such as the CT scanner **101** shown in FIG. 2 can use an alternating direction scanning C-arm or O-arm scanner geometry. Compared to a gantry-based configuration such scanner geometries provide a space-saving solution, but the rotation time is limited.

[0069] During the course of the intervention the CT scanner **101** runs in a continuously tomographic acquisition mode, while the image acquisition can be pulsed. Here, the first rotation can be used to run a fully sampled acquisition mode and all following rotations are performed in undersampled acquisition mode. Prior to the intervention, the prior image can for instance be sampled using a gantry-based system with a frame-rate of 30 frames per second, a rotation time of 10 s and a tube-current of 50 mA and tube voltage of 100 kV. During the course of the intervention, the temporal updates corresponding to undersampled sets of projections can be sampled with e.g. 18 frames per rotation, a tube current of 30 mA and a tube voltage of 100 kV. However, the scanning parameters may be adjusted according to the respective application.

[0070] During intervention the patient **106** is placed within the scanner system **101** and the information is provided to an operator, e.g. the interventionist **105**, through the array of displays **102**. Thereby the interventionist **105** stands next to the patient **106** and controls the intervention via the operator console **104**. In other embodiments, the interventionist **105** can also be situated in a remote location. The operator console **104** further allows modifying all functions of the interventional CT system **101** and most parameters affecting the imaging, e.g. the reconstruction algorithm, are controlled by the interventionist **105**.

[0071] During the intervention, a CT scanner system **101** acquires images of the patient **105**. Typically, such systems comprise a source **201**, **304** releasing electromagnetic radiation, preferably X-rays, and a detector **202**, **306** detecting the released X-rays after having traversed the objection to the image **106**, **302**, **310**, **203**. Thus, a typical result of such a measurement comprises projections of a three-dimensional (3D) energy distribution. In this sense, a projection is a two-dimensional (once integrated) or 1-dimensional (twice integrated) distribution of the underlying 3D energy distribution at specific projections solid angles of the detector with respect to the object to be imaged. Therefore, in order to reconstruct a 3D image of the object to be imaged **106**, **203**, **302**, **310** multiple projections are measured at different projection angles. From the multiple of projections a 3D image of the object to be imaged **106**, **203**, **310**, **302** can be reconstructed.

[0072] For measuring a multiple of projections, the CT scanner **101** typically comprises a gantry or C-arm-based construction for rotating the source **201**, **304** and the detector

202, 306 around the object to be imaged **106, 203, 302, 310**. FIG. 2 for example shows a C-arm-based CT imaging device with the source **201** and the detector **202** rotating around the object to be imaged, in this case the patient's head **203**.

[0073] Similarly, FIG. 3 illustrates a top view of the C-arm scanner system shown in FIG. 2. Both the source **201, 304** and the detector **202, 306**, which is preferably closed directly opposite to the source in one line through the object to be imaged **106, 203, 302, 310**. In order to acquire projections at different projection angles for reconstructing 3D images the source **201, 304** and the detector **202, 306** are rotated around the patient **203, 302, 310**. In FIGS. 2 and 3, the rotation direction is indicated by reference numerals **204, 311, 312**. In fact, the general CT acquisition process includes the source **201, 304**, and the detector **202, 306** rotating around the patient **203, 302, 310** while measuring line integrals along the X-ray direction **308**.

[0074] The algorithm performing the method according to the invention is implemented on the HPC **103**. During the course of the intervention, the HPC **103** calculates the updated image in real time. Here, standard CT density values and imaging features that are different from the standard CT values such as dual-energy index or difference images between actual projections and forward projected data sets are used to track instruments and to constrain the compressed sensing criterion. The updated image is then provided to the interventionist by displaying the updated image on the display array **102**. For display standard graphic volume display techniques, such as volume-rendering, surface-rendering or digitally reconstructed radiographs (DDR) are calculated from the volumetric dataset. The DDR are for example reconstructed from various angles which can depend on the radiologists selection or automatically depending on intrinsic imaging features so that the intervention guidance is optimized (e.g. perpendicular to the main movement direction of the catheter). Furthermore, angiographic features are incorporated into the DDR to provide a 3D road-mapping feature. All acquired projection images as well as all acquired temporal updates are stored in the HPC for later use, e.g. for a later reconstruction of bleedings and other modifications in soft tissue.

[0075] Additionally, the CT-scanner **101** can employ multiple X-ray sources as well as detectors up to arrays of X-ray sources combined with arrays of X-ray detectors. In such an embodiment different X-ray energies can be used and the instrument can comprise material that allows detection in dual-energy mode. Thus, characteristic absorption features of the instruments with respect to multiple radiation energy can provide further information on the instrument and its movement. This allows to track instruments using other means than standard CT absorption measurements and the detection of instruments is more robust.

[0076] FIG. 4.1 illustrates a flowchart **400** of the general CT workflow for 4D-CATH as proposed according to the present invention. In a first optional step, a number of projections fulfilling the Shannon-Nyquist criterion, i.e. a high number of projections is acquired at different projection angles. From these projections a tomographic reconstruction of a high quality prior image is performed. Herein the prior image serves as the first image of the object to be imaged. Furthermore, the instrument used during medical intervention on the object to be imaged is not present on the prior image. Therefore, the prior image comprises a volumetric image of the object to be imaged with high resolution prior to the interven-

tion. In other embodiments, such a first image may be provided from a database, a different imaging technique or a more coarsely sampled set of projections.

[0077] After acquisition and tomographic reconstruction of the first image, the instrument to be guided during the medical intervention on the object to be imaged is placed. Thus, in step **402** an instrument, such as a catheter, may be placed for further medical intervention, e.g. on the heart. After the placement of the instrument and on its way through the object to be imaged low dose update data is continuously acquired in step **403** for guiding the instrument during the intervention. Furthermore, the low dose update data from step **403** is continuously reconstructed using the imaging method according to the present invention.

[0078] Thus after a normal dose scan, all following scans can be performed as under sampled scans with a lower dose, which can be performed continuously. In practical interventional radiology, this results in an undersampling factor of the order of 10 to 30 resulting in 8 to 35 frames per reconstruction. These are reconstructed **404** providing images that represent temporal updates comprising changes in the examined volume. Update images are reconstructed using an iterative algorithm to incorporate the prior information as well as the actual temporal changes in the iterative steps. During the intervention the interventionist is continuously provided with updated images on-the-fly. Here the image diagnostically relevant for the physician is the sum of the prior image and the temporal change which is called the updated image.

[0079] FIG. 4.2 shows a flowchart **500** of the CT image acquisition and reconstruction. For the intervention in step **501a** high resolution, highly sampled CT scan is acquired as prior image. This image includes volumetric data representing the object to be imaged. During the intervention the prior image is used as a first image incorporated in the iterative construction algorithm during the intervention.

[0080] While performing the intervention by placing the instrument and moving it within the object to be imaged update information is continuously acquired in step **502**. The update information comprises undersampled sets of projections, which allow for a low-dose rate. In order to reliably reconstruct images for the physician performing the intervention, this update information is incorporated with the prior image in step **503**. Hence, the update information including the change of information corresponding to the moving instrument can be reconstructed in step **504** providing updated image data. Thus, the static part of the image is provided by the prior image, while the update information comprising a set of undersampled projections provides the temporal changes, which correspond to the moving instrument. This way a Prior Image Dynamic Computed Tomography (PRIDICT) may be realized allowing for real-time interventional guidance. Optionally, the projections measured during the intervention in step **502** may after reconstruction in step **504** be used to update the prior image. By the calculation of a new prior image any temporal changes occurring during the intervention, such as movement of patient, can be incorporated over time into the prior image leading to a higher image quality.

[0081] Furthermore, the update information comprising undersampled sets of projections may be collected in step **506** during the course of the intervention. After or during the intervention, but with a larger time delay than for **504**, the collected data sets from step **506** may be reconstructed in step **508** to visualize changes on a slower time scale than instru-

ment movements, such as bleedings. For this different reconstruction algorithms can be used, including algorithms with further sparsifying functions.

[0082] FIG. 5 shows an overview of the proposed reconstruction algorithm in relation to the amount of data that is acquired from the actual present (514). The continuous acquisition of low-dose tomographic data during the intervention 510 leads to a constantly increasing amount of projections from a time 514 (represents present) to a time 515 in the past. The longer ago from the present the higher the amount of low-dose tomographic data 512 that has been acquired. Thus at point 516.1, a short time after the present, the amount of tomographic data 512 only comprises a few projections. Longer time ago 516 the amount of low-dose tomographic data 512 increases by increasing the number of projections measured. At a point 516.2 in time, the number of projections measured during the intervention, i.e. the amount of low-dose tomographic data 512, is increased.

[0083] At the earlier point in time 516.1, the measured data is used 518.2 to reconstruct and display temporal changes, such as movement of guide wires or catheters 526. The reconstruction in step 521 can, e.g. be performed by using compressed sensing, where the sparsifying is done through a difference with e.g. the first image.

[0084] At a later point in time 516.2, the full amount of low-dose tomographic data 512 corresponding to the projections measured until then can be used 518.1 to visualize the anatomy, bleedings, bones, organs or other static data 524. Here, the reconstruction 520 can be performed using compressed sensing including further sparsifying functions such as gradient functions. The sparsifying functions can thereby be selected so that larger, more areal changes to the dataset will prevail, while shorter, more punctual changes will not be reconstructed. Furthermore, the data reconstructed after time period 516.2 may be fed back 528 into the reconstruction algorithm during intervention as a first image. Apart from the exemplary embodiments shown in FIG. 5, many more time points (such as 516.1 and 516.2) with different reconstruction algorithms, different sparsifying algorithms and different contributions to the displayed images exist and the person skilled in the art may adapt the explicit execution to the specific needs of the medical intervention.

[0085] FIG. 6 illustrates how undersampled sets of projections 604 collected during the intervention may be defragmented for creating a fully sampled set of projections 602 which again may be used as updated first image (sliding prior). For a sliding prior, individual projections of the undersampled sets of projections 606 are culminated in accordance with their solid angle 608. The projections of the update scans 606 are acquired at angular positions that are different from earlier projection positions so that after several rotations a new-fully sampled dataset is produced which can be used as a sliding prior. In the sliding prior, the changes (e.g. catheters, guidewires, etc.) are removed via standard image processing algorithms. Such algorithms can be similar to those being used in metal artifact reduction, where the reconstruction including high contrast signals is followed by the segmentation of high contrast signals and the elimination of these high contrast structures in projection data via forward projection of segmented high contrast data. The new reconstruction can then be performed without high contrast data. Alternatively or additionally, algorithms that track instruments can be used. There e.g. connected pixels in the data set can be identified, a comparison with a data base of possible instruments can be

performed or the PRIDICT (Prior Image Dynamic Computed Tomography) algorithm may even be modified so that significant pixels from the update will be memorized and removed. Other possibilities or additions may include dual-energy information e.g. through instruments, that provide a characteristic dual-energy signature.

[0086] FIG. 7 shows a flowchart 700 illustrating the general structure of the PRIDICT reconstruction algorithm that can be used for 4D-CATH. The proposed algorithm PRIDICT is a reconstruction technique specialized to interventional 3D and 4D applications. It incorporates the information of a former scan (prior scan) to the reconstruction process and can reduce the number of relevant measurements for temporal updates far under the Shannon-Nyquist sampling using the compressed sensing framework.

[0087] The interventional procedure starts with the acquisition of a fully sampled, normal dose scan 702 that can be used a prior image for the PRIDICT reconstruction algorithm as well as a first overview CT scan for the physician. This fully sampled scan can be reconstructed in step 704 through a standard CT reconstruction algorithm such as the FDK (Feldmann Davis Kress, as for example explained in Feldkamp L A, Davis L C, Kress J W. Practical cone-beam algorithm. J. Opt. Soc. Am. 1984; 1(6):612-619) to provide a prior image 706. The prior image 706 forms the basis image 708 for the iterative PRIDICT reconstruction algorithm.

[0088] During the intervention undersampled sets of projections 712 are measured. These provide the update information including static as well as dynamic components of the object to be imaged. In a first step of the algorithm, the image 708, which in the first iteration is equal to the prior image and includes volumetric data, is projected in accordance with the projection angles measured in the undersampled set of projections 712. The projected data 710 from the image 708 is then subtracted individually from the update projections of the undersampled set of projections 712 in operation 714.

[0089] The subtraction in operation 714 leads to difference images 716, which represent the difference between the undersampled set of projections and the projected prior image. These difference images 716 are reconstructed through a standard reconstruction routine in CT such as FDK to provide a reconstructed and fully volumetric difference image 718. In operation 720, the image 708, which in the first run corresponds to the prior image 706, is added to the volumetric difference image 718. In operation 722, image 706, which is the prior image and stays the prior image for every iteration, is subtracted. In the following steps 724, various image processing and mathematical operations, such as optimization routines, may be used to modify the image. This image is fed back into the iterative loop and serves as the base image 708 for the next iteration of the reconstruction algorithm.

[0090] In FIG. 8, the PRIDICT algorithm 800 is illustrated including a minimization loop in place of the mathematical operations performed in step 724 of FIG. 7. Equivalently to FIG. 7 a fully sampled set of projections 802 is measured, reconstructed via the FDK algorithm 804 to a volumetric data set 806, which is fed into the image 808 forming the basis of the PRIDICT algorithm. Then difference images 818 are reconstructed in 816 from the subtraction of the projections of the undersampled set of projections 812 and the projections projected from the image 808 in step 814. Then image 808 is added in step 820 following a subtraction of prior image 806 in step 822. Thus, the resulting image 824 excludes any static

components and highlights the change happened in the intervention during the measurement of the undersampled set of projections **812**.

[**0091**] In place of step **724** in FIG. **7**, the embodiment of FIG. **8** basically includes a minimization routing minimizing the number of significant pixels. The temporal updates are calculated as the FDK (Feldkamp-Davis-Kress) reconstruction of the difference of the actual measured projections and the calculated forward projections of the prior image. These FDK reconstructions contain only information of the current changes in the image but include a large number of streaking artifacts. To reduce these streaking artifacts, the total number of significant pixels (represented by the L0 norm) has to be minimized. Mathematically, the minimization of the L0 norm is difficult, so the L1 norm can be minimized alternatively, e.g. by using the method of the steepest gradient, other convex optimization techniques can also be used. Without constraints, the global minimum would be a zero matrix; however in practice this would imply no changes in the volume so that the prior image and the current image are identically. To eliminate the streaking artifacts without clearing the whole update, the minimization step has to be adjusted to the FDK reconstruction step, so that raw data congruence is aimed. This is presented in the next paragraph.

[**0092**] Without constraints, the global minimum of the L1 norm would be a zero matrix, but in fact this would eliminate any information in the update image. The link between the minimum number of independent probes and the number of significant pixels in the image: $m \approx S \ln(N)$ where $N \times N$ is the size of the reconstruction matrix, S is the number of significant pixels and m is the number of independent probes. Using this context, the maximum number of significant pixels in an image can be calculated for every given acquisition scenario. We call this context the CSC (compressed sensing criterion). The minimization process is continued as long as the CSC is not fulfilled. As soon as the L0 norm is smaller than the calculated maximum number of significant pixels, the CSC is reached, the minimization stops and the next iteration is performed. The L1 norm has not to be minimized directly, even optimizations minimizing L1 casually might be useful.

[**0093**] As described, L1 is minimized because of the mathematical difficulties minimizing L0. The actual aim is to minimize L0, so in an embodiment of the algorithm, other optimizations minimizing L0 casually or directly may be used, even if they do not minimize L1.

[**0094**] Within the minimization loop after operation **822**, the signum of the image **824** including only the temporal changes is calculated in **826** and subtracted from the image **824** in operation **828**. From there it can be checked whether the compressed sensing criterion **830** is fulfilled. If it is fulfilled, the image is fed back into **808** replacing this image and image **808** can be displayed to the operator. If the compressed sensing criterion in **830** is not fulfilled, the image will be fed back into the compressed sensing minimization loop to **824**.

[**0095**] In contrast to the reconstruction algorithm shown in FIG. **8**, FIG. **9** includes a further sparsifying transform **936** performed before the minimization of the number of significant pixels. As sparsifying transform **936** gradient operations, wavelet transformations, curvelet transformations, contourlet transformations or a combination thereof can be used. This transform is then applied after reconstructing the difference **918** and before the minimization loop is entered in **924**.

[**0096**] In step **936** it is also possible to use a combination of different transforms and/or tunable transformations, which

can be used in different configurations during one reconstruction, to correctly reconstruct different structures point-like or curve-like structures. Furthermore, the weight of different sparsifying transforms can be varied e.g. according to the sparseness of the transformed image.

[**0097**] The realization of such a tunable PRIDICT reconstruction algorithm **1000** is shown in FIG. **10.1**. The algorithm shown in FIG. **10.1** corresponds to the algorithm shown in FIGS. **7** to **9**. The embodiment of FIG. **10**, however, incorporates the image processing and mathematical operations **1024** including the minimization loop **1028** for finding an optimal difference image and a pre-applied sparsifying function **1026**. Here, the sparsifying function as well as the minimization loop **1028** may be influenced by the actual difference **1030** given in the static prior set of projection **1002** and the undersampled set of projections **1012** including the temporal changes. So certain parameters of the mathematical functions, i.e. the sparsifying function and the minimization, may be modified in accordance with a comparison **1030** done between the prior set of projections **1002** and the undersampled set of projections **1012**. Thus, the reconstruction algorithm PRIDICT **1000** can be tuned and adapt to different situations depending on the temporal changes. This is particularly advantageous when reconstructing temporal changes from guide wires, catheters, tubes and the like or temporal changes which move much slower such as bleedings.

[**0098**] FIG. **10.2** shows further possibilities for influencing the reconstruction of updated images. The flowchart **1040** comprises in step **1042** the acquisition of undersampled imaging data during the intervention. From this data and the previously acquired prior image or an updated prior image the amount of difference and/or movement in the examined volume is calculated in step **1044**. Then the calculated amount of difference and/or movement can be used to influence scan parameters and/or reconstruction parameters, such as the number of projections included into the undersampled set of projections, in step **1046**. Optionally, such influences can also be triggered from the outside such as an operator **1052**. Furthermore, these influences are used to vary the reconstruction algorithm **1048**, such as reconstruction matrix size, time resolution, sparsifying function and so on. Lastly, a marker for the degree of completeness of the imaging data may be provided in operation **1050**.

[**0099**] In order to provide a parameter of completeness, FIG. **11** shows another realization of the PRIDICT reconstruction algorithm **1100**, which broadly corresponds to the reconstruction algorithm PRIDICT shown in FIGS. **7** to **9**. Here the subtraction **1114** is not only used to calculate the difference image **1116** but also to calculate the total modification **1120** in the image. From there, the matrix size can be set **1122** which can be fed back to the difference reconstruction image **1118**. This results in optimized reconstruction minimization of the significant number of pixels. Thus, if there is a large change the modifications are large and the matrix size may be smaller. On the other hand, if the changes between the prior image data and the undersampled set of projections are small, the modifications are small and the matrix size may be set to a larger value in order to reconstruct the difference properly.

[**0100**] FIG. **12** illustrates the effect of the proposed PRIDICT reconstruction algorithm on a head. A pig head was scanned prior to the intervention, the result of which is shown in FIG. **12.1**. After the insertion of a guide wire into an arteria

a temporal update was provided. Here, FIG. 12.2 shows the reconstruction of the undersampled set of projections using a standard FDK algorithm. In comparison to FIG. 12.1 showing the result reconstructed by a standard FDK algorithm of the fully sampled prior image, FIG. 12.2 includes streaks and artifacts. In FIG. 12.3, the result including the guide wire is shown reconstructed through the PRIDICT reconstruction that uses a fully sampled prior image and an undersampled set of projections. As can be seen, the result of FIG. 12.3 is artifact-free and the magnified inset shows the wire labeled through the smaller arrow and the vessel through the larger arrow.

[0101] FIG. 13 illustrates different embodiments 1302, 1306, 1304 for motion compensated reconstruction in connection with the proposed PRIDICT reconstruction algorithm. In the first column 1300 of FIG. 13, a prior image 1308 reconstructed from a fully sampled set of projections 1312 includes different phases 1310 of the cardiac motion. As a consequence of the cardiac motion, its appearance changes from projection to projection. Therefore, the prior image 1308 results in a smeared image of the heart 1322, which is smeared by the heart motion during the measurement. The bottom part of column 1300 illustrates the corresponding undersampled set of projections where different projections represent different cardiac phases 1318 as well. Here again the heart 1318 periodically changes appearance for the different cardiac phases 1318 from projection to projection 1316.

[0102] Column 1302 illustrates one implementation for motion-gated reconstruction. Under the assumption that the heart appearance is the same for a single cardiac phase over the entire scan time, a gating is performed to assort the acquisition into the heart cycle which itself is divided into phase bins with sufficiently small widths. Thus, the displacement of the heart is taken into account for e.g. each projection measured in the undersampled set of projections. In the example illustrated in column 1302, bottom part, the undersampled set of projections 1316 is binned into three different phases of the heart beat cycle 1324, 1326, 1327. These phases may for instance be monitored throughout the intervention, e.g. through electrocardiography, and the binning is carried out in accordance with the monitored reference signal. Incorporating the prior image 1322 using the PRIDICT algorithm to reconstruct the heart in each individual phase 1324, 1326, 1327 results in update images which show the instrument 1313 in the equivalent place for each cardiac phase 1324, 1326, 1327. This way, the smearing due to the heart motion can be compensated for and the temporal change due to the instrument 1313 can be reconstructed according to the cardiac phases 1324, 1326, 1327. Thus, the interventionist can at each point during the intervention assess where within the heart the instrument is situated. Owing to the reduced smearing resulting from the cardiac motion a more accurate position of the instrument 1313 within the heart can be displayed to the operator. The advantage of such an algorithm is that there is no gating signal necessary for the prior image.

[0103] In column 1306 another implementation of motion-compensated reconstruction via PRIDICT is illustrated. In this case the prior image 1308 is binned into the cardiac phases 1334, 1336, 1338 rather than the reconstructed update image 1340. Here no gating signal is necessary for the reconstruction of the heart phases and thus, slower scanner systems might be utilized.

[0104] Column 1304 shows another implementation of motion-gated reconstruction through PRIDICT. Here, rather than reconstructing the time frames with respect to cardiac and/or respiratory phases or the prior as shown in columns 1302, 1306, the prior as 1308 well as the time frames 1314 may be reconstructed with respect to the cardiac and/or respiratory phase using compressed sensing reconstruction. In this embodiment, gating is necessary for the prior as well as the update image and images can be reconstructed with less motion-related smearing.

[0105] Furthermore, the reconstruction scan can be incorporated with low dose update scans using motion-compensated reconstruction combined with compressed sensing and taking a 4D representation of the moving heart into consideration. The idea of motion-compensated 4D reconstruction may also be deeply integrated into the reconstruction algorithm. In order to do so, the cardiac and/or respiratory phases are registered or the transformation may be done through morphing or movement field. By using a transformation the image may be projected into either the moving space or a static space. In a static space the object to be imaged may be displayed in one phase only, which is particularly useful for the guidance of catheters. Furthermore, the requirements to the scanning speed are relaxed. With appropriate motion-compensating reconstruction algorithms (including movement vector fields) the data that is acquired at a certain heart phase can be used to reconstruct images at a different heart phase.

[0106] FIG. 14 shows an overview illustrating the capability of the PRIDICT reconstruction algorithm. Starting from a full dose prior image 1410 of the object to be imaged during intervention with a moving instrument low dose updates 1412 are acquired. From these low dose updates a 4D data set of the moving instrument 1414 may be reconstructed. Furthermore, the low dose updates 1418 may be used to reconstruct the 3D data set of a body part with intervascular contrast media 1420. Lastly, the 4D data set of the moving instrument 1414 and the 3D data set of the vascular structure 1420 can be merged to a 4D intervention guidance in 3D road-mapping 1416. Thus, the movement of the instrument during the intervention may be visualized after the intervention in the form of a roadmap illustrating the full course of the intervention.

[0107] The present invention relates to an imaging message for radiologically guiding an instrument during medical interventions on an object using X-rays. The invention further refers to a system for radiologically guiding medical interventions on an object according to the proposed method. Without limiting the foregoing, further embodiments are briefly lined out in the following:

[0108] In one embodiment, the method for performing imaging during radiological interventions, comprising the steps

[0109] Measurement of more than one set of projections of an object to be imaged at different points in time, wherein the measurement is performed by an imaging apparatus,

[0110] Dynamic reconstruction of volumetric datasets from the more than one set of projections, wherein the reconstruction is performed by processing means.

[0111] The method according to the present invention allows for dynamic imaging. In dynamic imaging, volumes are acquired in close, timely consecution. Here the term timely is used in the sense temporal or equivalently in the sense of different points in time. Volumetric describes that image data sets are reconstructed in 3 spatial dimensions. In

combination dynamic volumetric imaging describes 3D+time imaging or 4D imaging respectively. Dynamic imaging thus allows for controlling and monitoring interventions in three spatial dimensions. Such interventions are preferably performed on body parts, such as the cardiovascular system, tubular organ structures or on the brain, which exhibit complicated three dimensional structures. Furthermore, the time component, i.e. the dynamics, of interventions, such as in catheter interventions, bronchoscopy interventions, implantation of cardiac pacemakers or positioning of stents, plays an important role in order assure save performance interventions. With the proposed method for dynamic imaging temporal changes in the object to be imaged, such as instrument movements, can be controlled and monitored during the intervention. This provides the interventionalist with a high degree of certainty in the way the intervention is performed and thus reduces the risk for the patient during intervention. In particular maneuvering in complicated three dimensional structures becomes easier and misplacement of e.g. stents or injuries due to e.g. rupture of vessels can be avoided. Overall the imaging method according to the present invention provides full control in three spatial dimensions including temporal changes during interventions.

[0112] In contrast to prior known intervention guidance techniques, which are not 4D and leave the interventionalist with certain degree of uncertainty regarding the position of the instruments and the patient condition, the systems and methods of the present application embody a method of 4D intervention guidance (named 4D-CATH which stands for 4D Catheter Advancement with Tomographic Help).

[0113] In one embodiment of the invention, the imaging is a tomographic system, such as a magnetic resonance imaging (MRI) scanner or a computed tomography (CT) scanner. In the case of MRI non-ionizing radio frequency (RF) signals are used to acquire images. CT, on the other hand, uses X-rays (a type of ionizing radiation) in order to image objects. Owing to the dose X-rays carry, only a restricted number of exposures can be performed and the number of projections to be measured is to be kept at a minimum. Preferably, the imaging system utilized in connection with the present invention is a flat-panel cone-beam CT system and a CT reconstruction method to calculate 3D data of the examination volume.

[0114] The examination volume is the volume that is being imaged or any part of it. The maximum size of the examination volume is limited due to the field of measurement of the tomographic system. Examination volume can include the patient and instruments in it. The examination volume is defined by physical characteristics of the scanner system.

[0115] In one embodiment the CT data (CT fluoroscopy) are continuously acquired to provide 4D information. The systems and methods of the present application as applied to intervention guidance provide 4D imaging during interventions. Some embodiments reduce the radiation dose during CT fluoroscopy, reduce imaging artifacts and provide the radiologist with 4D imaging data.

[0116] Exemplary embodiments of the present application can comprise a CT reconstruction algorithm based on compressed sensing. Or the reconstruction can be based on McKinnon-Bates, PICCS or TRI-PICCS. Furthermore, in one embodiment the underlying CT reconstruction algorithm is based on prior knowledge of the scanned volume.

[0117] Certain embodiments employ the prior knowledge of a prior data set that is updated with few CT projections during intervention. The update is preferably derived with a

few projections or very low-dose projections. The reconstruction to combine a prior scan and the update information can preferably be an iterative CT reconstruction method, which can be based on the theory of compressed sensing.

[0118] The prior data and update information can be acquired over a limited angle orbit around the patient. The reconstruction algorithm can use information that it derives by comparing the update scan with the prior scan to correct for limited angle CT reconstruction artifacts or distortions. This can be done by calculating a local distortion parameter by comparing the sparsely sampled limited angle updates with the well sampled prior scan.

[0119] In one embodiment of the invention the continuously acquired data sets are collected and reconstructed from data collected throughout various amount of rotations, sampling rates and hence with a different time resolution and image qualities (SNR, soft-tissue contrast, etc.). The position of the projection positions might vary between rotations to acquire highly sampled data sets from several rotations.

[0120] In one embodiment the system provides the radiologist with information about the degree of completeness of the data. In one embodiment this might be realized through variations in the reconstruction matrix size.

[0121] One embodiment includes a system to track the interventionalists hands and avoid the radiation of the hands when they cross the direct X-ray beam area.

[0122] One embodiment employs a registration algorithm that compensates for movements of the patient or system between acquisition of the prior scan and the update scans. In one embodiment a new prior image is constantly updated from the projection data.

[0123] One embodiment includes iterative reconstruction algorithms, where the result of a former reconstruction (f-1) is used as the initial input of the actual reconstruction (f). Alternatively, an assumed three dimensional structure can be used as a first guess.

[0124] In one embodiment the changes within the examined volumes are measured during intervention and the amount of changes influence the scan and imaging parameters (e.g. the number of projections used for the temporal update, the temporal resolution, the number of reconstructions shown to the interventionalist, the used sparsifying function, the X-ray tube current, the voltage, etc.). The amount of changes is a parameter proportional to the significant pixels; this can be the number of pixels that are different in the prior image and the temporal update. However, the amount of changes can also be proportional to the number of significant pixel in any domain that was used by the sparsifying function. In one version the amount of changes are measured by comparing projections of the PRIOR with projections of the update scan; in one version of the embodiment, the amount of changes between consecutive scans are measured by calculating the difference images, in one version measured projections are compared with forward-projected projections. In one embodiment the amount of dose that is applied will be influenced by the amount of difference in the timely updates. In one embodiment the radiologist will get information about the degree of completeness of the reconstructed scan data. The uncertainty in the information will be expressed through a display or uncertainty in the reconstruction dataset.

[0125] In one embodiment the algorithm incorporates information about the position of vessels from angiography in the reconstruction process. In one embodiment this is used to

calculate a likelihood of vessel perforation, to even further reduce the amount of necessary projections for guide-wire guidance by assuming the most likely position of instruments will be within vessels.

[0126] In one embodiment multiple X-ray sources work at different energies. Potential material differentiation is being used to influence the reconstruction algorithm. In one embodiment the dual-energy information is being used to which information changes are related to instrument movements.

[0127] In one embodiment 4D digital subtraction angiographies are reconstructed to provide a 3D road map for intervention guidance. In one embodiment the 4D angiography is reconstructed by combining a highly-sampled prior scan without contrast media with sparsely-sampled updates that are acquired during the contrast media injection.

[0128] Yet another embodiment relates to a computer readable medium having recorded instructions which, if executed by a computer, would cause a method to be performed. Such a method comprises (a) capturing a plurality of images; (b) optionally acquiring a prior imaging and timely updates during radiological interventions; and (c) combining the timely updates with the PRIOR, potentially using iterative CT reconstruction methods potentially based on compressed sensing.

[0129] Both the foregoing summary and the following brief description of the drawing and the detailed description are exemplary and explanatory and are intended to provide further details of the compositions and methods as claimed. Other objects, advantages, and novel features will be readily apparent to those skilled in the art from the following detailed description.

[0130] FIG. 1 shows an exemplarily configuration of the 4D-Cath-Lab with a closed gantry for continuous rotations (101), an array of displays (102), both connected to a high performance computing setup (103). All functions of the interventional CT systems and most parameters affecting the reconstruction can be controlled using the operator console (104). The position where the interventionist stands is marked as 105.

[0131] FIG. 2 is an illustration of the image acquisition phase of a dynamic interventional C-arm CT imaging device with source (201) and detector (202) are rotating around the patient (203).

[0132] FIG. 3 is an illustration of the general CT acquisition process with a circular trajectory (302), the source (304) and detector (306) rotating around the patient (310) while measuring line integrals along the X-ray direction (308).

[0133] FIG. 4.1 is the flowchart of the general CT workflow for 4D-CATH.

[0134] FIG. 4.2 is a flowchart of CT image acquisition and reconstruction.

[0135] FIG. 7 is a flowchart showing the general structure of the PRIDICT reconstruction algorithm that can be used in 4D-CATH.

[0136] FIG. 8-11 are potential variations of the PRIDICT algorithm.

[0137] FIG. 10.1 is an example of a method to influence the sparsifying function and the update reduction function depending on the amount of differences in the update and prior projections.

[0138] FIG. 5 illustrates one of the possible modifications to the proposed algorithm collecting projection and reconstruction datasets for a delayed reconstruction of soft tissue contrast.

[0139] FIG. 10.2 is a flowchart showing a modification of the algorithm in which the continuous data acquisition process and image reconstruction algorithm is varied with respect to the actual differences/movements in the acquired volume.

[0140] FIG. 10.2 Workflow to integrate the radiologists' input to influence the degree of changes to the image acquisition and reconstruction algorithm according to the amount of changes in the examined volume.

[0141] FIG. 12 shows exemplarily reconstructions of a guide wire moved in a pig's head. Fully sampled standard reconstruction is compared to the undersampled standard reconstruction as well as to the PRIDICT reconstruction using fully sampled prior and undersampled temporal update. The prior is a fully sampled dataset reconstructed FDK (Feldkamp-Davis-Kress) algorithm (902). The update is a factor 20 under-sampled CT dataset of the same volume, but with the guide wire inserted, standard FDK reconstruction leads to 904, reconstruction of prior together with update using PRIDICT results in 906.

[0142] One of the preferred embodiments of the algorithm is a gantry-based CT scanner system with a flat detector-based imaging chain (FIG. 1). Another embodiment consists of a C-arm or O-arm scanner geometry (FIG. 2).

[0143] The algorithm can in general be used with any kind of imaging modality with a continuous tomographic acquisition.

[0144] A continuously rotating CT gantry is the preferred implementation but e.g. alternating direction scanning C-arm systems can be used as well. Furthermore, various scanning trajectories can be incorporated. Additionally, multiple X-ray sources as well as detectors up to array X-ray sources combined with array X-ray detectors could be employed.

[0145] The system can be run in a continuously tomographic acquisition mode and the image acquisition can be pulsed. In a potential implementation using a continuously rotating CT gantry, the first rotation would be run in full sampled acquisition; all following rotations are performed with a fraction of projections acquired.

[0146] The tomography system can be directly connected to a high performance computing setup (HPC) like conventional clusters, GPU-systems, GPU-clusters, cloud systems or other mainframes.

[0147] The algorithm can be implemented into the HPC system; the system calculates the current image to be shown to the interventionist and displays it in various ways on the display array. This can happen in real-time. Standard graphic volume display techniques can be employed, such as volume-rendering, surface-rendering, etc. In one embodiment DDR (digitally reconstructed radiographs) are calculated from the volume dataset. In one embodiment the DDR are reconstructed from various angles which can depend on the radiologists selection or automatically depending on intrinsic imaging features so that the intervention guidance is optimized (e.g. perpendicular to the main movement direction of the catheter). In one embodiment angiographic features are incorporated into the DDR to provide a 3D road-mapping feature.

[0148] All acquired projection images as well as all acquired temporal updates may be stored in the HPC for later use, e.g. for a later reconstruction of bleedings and other modifications in soft tissue.

[0149] The proposed Algorithm PRIDICT (Prior Image Dynamic Computed Tomography) (FIG. 7 or 8) is a recon-

struction technique specialized to interventional 3D and 4D applications. It incorporates the information of a former scan (prior image) to the reconstruction process and can reduce the number of relevant measurements for temporal updates far under the Shannon-Nyquist sampling using the compressed sensing framework.

[0150] The prior scan may be fully sampled, meaning the number of measured projections comply with the Shannon-Nyquist theorem, while the temporal updates are highly undersampled and do not satisfy the Shannon-Nyquist theorem. In the following, the algorithm is described in three steps:

[0151] Image acquisition and processing including the reconstruction are shown in FIG. 4.1, 4.2, 5. The interventional procedure starts with the acquisition of a fully sampled, normal dose scan that can be used as prior image for the PRIDICT reconstruction algorithm as well as first overview CT scan for the physician (501). After that normal dose scan, all following scans can be performed as undersampled scans with a lower dose (403), which can follow different trajectories (examples in FIG. 12). In practical interventional radiology, this could be an undersampling factor in the order of 10-30 resulting in 8-35 frames per reconstruction. These reconstructions are called temporal updates (403), because they represent only the changes in the examined volume. These update images are reconstructed using an iterative algorithm to incorporate the prior information as well as the actual temporal information in every iterative step (503). The image diagnostically relevant for the physician is the sum of the prior image and the temporal update which is called current image.

[0152] The raw data-based image reconstruction can be divided in two separate categories, the reconstruction of the prior image and the reconstruction of the temporal updates.

[0153] The reconstruction method of the prior image in generally is independent of the rest of the algorithm, every algorithm can be used assuming that the reconstructed prior image is free of artifacts and not dominated by noise.

[0154] The temporal updates are calculated as the FDK (Feldkamp-Davis-Kress) reconstruction of the difference of the actual measured projections and the calculated forward projections of the prior image. These FDK reconstructions contain only information of the current changes in the image but include a large number of streaking artifacts. To reduce these streaking artifacts, the total number of significant pixels (represented by the L0 norm) has to be minimized. Mathematically, the minimization of the L0 norm is difficult, so the L1 norm can be minimized alternatively, e.g. by using the method of the steepest gradient, other convex optimization techniques can also be used. Without constraints, the global minimum would be a zero matrix; however in practice this would imply no changes in the volume so that the prior image and the current image are identically. To eliminate the streaking artifacts without clearing the whole update, the minimization step has to be adjusted to the FDK reconstruction step, so that raw data congruence is aimed. This is presented in the next paragraph.

[0155] Regularization of the minimization process. Without constraints, the global minimum of the L1 norm would be a zero matrix, but in fact this would eliminate any information in the update image. The link between the minimum number of independent probes and the number of significant pixels in the image: $m \approx S \ln(N)$ where $N \times N$ is the size of the reconstruction matrix, S is the number of significant pixels and m is

the number of independent probes. Using this context, the maximum number of significant pixels in an image can be calculated for every given acquisition scenario. We call this context the CSC (compressed sensing criterion). The minimization process is continued as long as the CSC is not fulfilled. As soon as the L0 norm is smaller than the calculated maximum number of significant pixels, the CSC is reached, the minimization stops and the next iteration is performed.

[0156] The L1 norm may not be minimized directly, even optimizations minimizing L1 casually might be useful.

[0157] As described, L1 is minimized because of the mathematical difficulties minimizing L0. The actual aim is to minimize L0, so in an embodiment of the algorithm, other optimizations minimizing L0 casually or directly may be used, even if they do not minimize L1.

[0158] In one embodiment of the algorithm, wavelets, curvelets or contourlets can be used for the sparsifying transform. In embodiment the sparsifying function might be applied after reconstruct the differences (FIG. 9).

[0159] In another embodiment of the algorithm, a combination of different transforms may be used and/or one tunable transformation can be used in different configurations during one reconstruction to correctly reconstruct different structures as point-like or curve-like structures.

[0160] The weight of different sparsifying transforms can be varied e.g. according to the sparseness of the transformed image.

[0161] Accordingly, one method of the present application entails guidance of catheters, tubes, syringes, needles, guide-wires, coils, stents, vessel prosthesis and similar devices. Similarly intervention devices such as biopsy needles, drainage needles, catheters, thermo ablation and chemoembolization devices are imaged. Potentially visible embolization agents are visualized as well as contrast media of different kind.

[0162] A combination of all acquired projections or a part of them can be used for a reconstruction of soft tissue images and/or changes, intracranial hemorrhage etc. This feature can be used for an intra-operative detection of possibly vessel damages. Therefore, the single temporal updates can be reprojected. Out of measured and calculated projections, a stationary projection without any moving structures can be calculated (e.g. by subtracting moved parts from the measured projections).

[0163] A function to adopt the spatial resolution in the image (the image matrix) to the number of samples and the changes in the examined volume can be included. A method to calculate the changes and the sparsity in the used image domain is required, one possibility for that can be sum of squared differences of measured update projections and calculated forward projections (FIG. 11).

[0164] Preferred workflow looks as follows: The patient is placed in a continuously rotating CT gantry and scanned once with high dose and full sampling when the surgery starts. While the physician moves the guide wires and catheters, the CT system acquires undersampled CT datasets. The result of the reconstruction using a PRIDICT-based algorithm is displayed directly on the displays in different visualizations (FIG. 4.2).

[0165] FIG. 3 schematically depicts an exemplary imaging apparatus 300. Imaging apparatus 300 has a gantry 302 to carry source and detector devices 304 and 306, with a hollow center to allow the insertion of a test subject 310. Rotatable around the gantry 302 in the present example is a source 304,

which can be, for example, an X-ray source for X-ray CT imaging. Source 304 produces radiation which travels in exemplary fashion along path 308, to reach detector 306. The radiation passes through subject 310. Thus, detector 306 records the intensity of the transmitted radiation.

[0166] Both source 304 and detector 306 may be rotated about gantry 302 to capture a plurality of images at varying angles θ , which may be measured from any cross-sectional line through the hollow 302. To be rotated about means to be positioned at more than one angle θ along, allowing the acquisition of two or more images at different angles.

[0167] Imaging apparatus 300 may preferably be a CT scanner (e.g. VCT, Siemens Medical Solutions, Forchheim, Germany), which includes a flat detector and a modified X-ray tube both mounted on a multi-slice CT gantry. The flat detector (PaxScan 4030CB, Varian Medical Systems, Palo Alto, Calif.) employed by this scanner consists of 2048x1536 detector pixels on an active area of 40x30 cm² resulting in a pixel size of 1942 μm^2 . X-ray photons are converted to light photons in a scintillation layer of thallium-doped caesium-iodine crystals which are subsequently detected by the photodiodes of the amorphous silicon wafer.

[0168] The X-ray tube may be modified to increase the anode angle to increase the cone angle of the generated X-ray beam. The focal spot size can be reduced to 0.5 mm to decrease the effect of penumbral blurring. An even smaller focal spot is technically feasible. This may result in a reduced photon flux and could make advisable longer exposure times.

[0169] Taking the geometry of the present example of the scanner into consideration, the scanner's total scan field-of-view is 25x25x18 cm³. The active detector area may be reduced to 192 lines in z-direction and 1024 rows in x-y-direction to increase frame rate. The detector can be read out in a 2x2 binning mode, meaning that four neighbouring pixels are averaged. This results in a decreased scan field-of-view of 25x25x4.5 cm³, which is still large enough to cover the entire thorax and diaphragm of a rat. The resulting frame rate is 100 frames per second (fps), which translates into an exposure time of 10 ms per projection.

[0170] The spatial resolution of the preferred scanner mentioned above, as computed by scanning a tungsten wire phantom, is 24 lp/cm at 10% modulation transfer function. This isotropic spatial resolution translates into a minimal detectable feature size of 200 μm . Each projection image acquired during gantry rotation is time stamped and is labeled with the angle of acquisition. Scanner rotation times can be varied from 2 s to 19 s in steps of 1 s, the maximum total scan time being 80 s. The tomographic image reconstruction is based on a modified Feldkamp algorithm.

[0171] A tube voltage of 80 kV and a tube current of 50 mA with continuous radiation may preferably be selected.

[0172] Various aspects of the present application will be illustrated through examples depicted in FIGS. 3-8. In FIG. 4, an overall method for generating intervention guidance image data according to the present application is depicted.

[0173] In a preferred embodiment, continuous rotations are carried out, with one rotation being performed in 5 seconds. One hundred projections are generated per second, and each rotation is a full 360 degrees. Thus, there can be 500 projections per rotation, and 8000 per total imaging time. The amount of projections acquired can be substantially reduced. During intervention guidance, the prior is acquired with 500 projections and for 4D intervention guidance a 2 second

rotation time is used, together with 20 frames per second. Using PRIDICT or a variation of it, the continuous updates will be reconstructed.

[0174] The effect of the PRIDICT reconstruction algorithm is visualized in FIG. 9. A pig corpus' head was scanned prior the intervention and after the insertion of a guide wire into an arteria as a temporal update. The reconstruction of the fully sampled (902) and undersampled (904) temporal updates reconstructed using a standard FDK algorithm are shown. In contrast to the undersampled reconstruction 904 that is dominated by streaking artifacts, the PRIDICT reconstruction 906 that uses a fully sampled prior image and an undersampled temporal update is artifact free.

[0175] It is also possible that the angular positions within each rotation are not identical to angular positions within other rotations. This can be used to reconstruct more complete data sets (e.g. with a good soft-tissue contrast) by combining information from several rotations (FIG. 7).

[0176] The imaging system could be able to provide real-time updates of the imaged volume including instrument positions during interventions. In one embodiment the images are displayed in volume-rendering, surface rendering or such alike. 3D subtraction angiography can also be performed by subtracting projections or reconstructed volumes with or without contrast media. Furthermore, a 3D road-mapping feature can be integrated, here the contrasted vessels are superimposed in 3D on timely updates to guide the intervention. Furthermore, information of the vessel positions can be integrated into the reconstruction algorithm and can be used to manipulate the sparsifying function.

[0177] The system can be built with several imaging chains, X-ray tubes and detectors. The angular distance between X-ray tubes can be 90°, for example. Different X-ray and detector systems can be employed, also in one system. The tube voltage can be varied to provide multi-energy information. The multi-energy information can be used to constrain the prior images, to detect changes in the volume (e.g. the different expression of guide-wire and tubing in multi-energy systems). The system can provide methods to vary the angular distance between projections and the angular distance over which one projection is acquired. This can be through a very fast x-ray on/off method, same kind of shutter or a frame work installed within the gantry.

[0178] The system can be built so that it can provide all relevant features of current standard angiography C-arm system, which are X-ray fluoroscopy and digital subtraction angiography. In one embodiment this can be done by parking the imaging chain at a certain angular position.

[0179] In one variation of 4D-CATH the data acquisition is varied according to the amount of changes or movements in the examined volume. In one embodiment the differences of projections of several rotations are compared. In case of little differences the projection acquisition rate or radiation dose is reduced (FIG. 8.). This can also be used to compensate patient movements that would yield great differences in the acquired projection data and hence require more radiation dose to compensate for it. This feature could be represented within the reconstruction algorithm as described in FIG. 6/3. In one embodiment the radiologist can influence the effects that the amount of differences has on the reconstruction algorithm or data acquisition (FIG. 9). E.g. by selecting certain programs or modes of the system and by pressing a handle or a pedal the radiation dose will be much more increased with the same amount of changes in the examined volume.

[0180] In one embodiment the prior scan will be continuously updated by the continuous data acquisition. In one current realization the prior will be generated out of the last 10 rotation consisting of projections that are acquired at slightly different angular positions.

[0181] In one embodiment a spatial-registration algorithm is incorporated that registers the prior scan data with the update scan data using standard registration methods based in the projection or volume domain.

[0182] The PRIDICT algorithm can be implemented in various ways using different sparsifying transforms. One of the simplest implementations is the use of the difference as sparsifying transform. Other possible transforms are e.g. the calculation of the gradient, a wavelet-sparsifying or a sparsification using dual energy concepts. As an alternative to one fixed sparsifying function, a tunable function can be used representing different aspects of interventional tools (FIG. 9).

[0183] Although embodiments of the present invention have been described with reference to particular examples, it will be clear to a person of skill in the art based on the teachings herein that certain modifications can be made in the described examples. Throughout the specification, any and all references to a publicly available document, including a U.S. patent, are specifically incorporated by reference.

[0184] In one embodiment of the invention the method for performing imaging during radiological interventions, comprises the steps of measuring more than one set of projections of an object to be imaged at different points in time, wherein the measurement is performed by an imaging apparatus and dynamic reconstruction of volumetric datasets from the more than one set of projections, wherein the reconstruction is performed by processing means. In a further embodiment of the invention, the more than one set of projections of the object to be imaged are measured in a repetitive or continuous scanning mode of the imaging apparatus. In a further embodiment of the invention, the processing means comprise a processor. In yet a further embodiment of the invention, the more than one set of projections of the object to be imaged comprise undersampled sets of projections. In a further embodiment of the invention, the undersampled sets of projections are measured at consecutive points in time during the radiological intervention. In a further embodiment of the invention, the more than one set of projections of the object to be imaged comprises at least one fully sampled set of projections, preferably measured before, during or after the radiological intervention. In a further embodiment of the invention, the reconstruction is performed by an iterative reconstruction method. In a further embodiment of the invention, the iterative reconstruction method is based on compressed sensing theory. In a further embodiment of the invention, the variations in an algorithm for the reconstruction are matrix size, interruption criterion, sparsifying functions. In a further embodiment of the invention, the interruption parameter of the iterative reconstruction method is depending on the amount of changes in the volumetric datasets reconstructed from the undersampled sets of projections, the amount of significant pixels and/or the used sparsifying function. In a further embodiment of the invention, the more than one set of projections of the object to be imaged comprise at least one fully sampled set of projections and undersampled sets of projections measured at consecutive points in time during the radiological intervention, wherein the reconstruction is configured to combine the at least one fully sampled set of projections with undersampled sets of projections. In a further

embodiment of the invention, imaging parameters of the imaging apparatus depend on the amount of movement and information changes in an examination volume. In a further embodiment of the invention, the dependency on the amount of movement and information changes in the examination volume is influenced by the interventionalist. In a further embodiment of the invention, the step of the reconstruction is influenced by the amount of changes in the object to be imaged. In a further embodiment of the invention, a user is provided with some means to influencing the ratio how changes in the examined volume influence the data acquisition, reconstruction parameters or data display.

[0185] In another aspect of the invention, the method described above is for use during radiologically guided interventions on the cardiovascular system. In a further aspect of the invention, the method described above is for use in the implantation of cardiac pacemakers. In a further aspect of the invention, the method described above is for use during radiologically guided interventions on tubular organ structures, preferably lungs or kidneys. In a further embodiment of the invention, for use in positioning of stents in vessels or bronchi. In a further aspect of the invention, the method described above is for use during bronchoscopy interventions. In a further aspect of the invention, the method described above is for use during catheter interventions. In a further aspect of the invention, the method described above is for use during radiologically guided interventions on the brain.

[0186] In one embodiment of the invention a system for carrying out the method as described above, contains an imaging apparatus in communication with processing means, wherein the imaging apparatus is a tomographic system, such as a magnetic resonance imaging (MRI) scanner or a computed tomography (CT) scanner. In another embodiment of the invention the computed tomography scanner comprises at least one X-ray source and at least one detector, wherein the X-ray sources differ in terms of the X-ray spectra or the detectors providing means of energy differentiation.

LIST OF REFERENCE NUMERALS

[0187]	100	Scan system
[0188]	101	CT scanner
[0189]	102	Array of displays
[0190]	103	HPC
[0191]	104	Operator control
[0192]	105	Operator
[0193]	106	Patient
[0194]	201	Source
[0195]	202	Detector
[0196]	203	Object to be imaged
[0197]	204	Rotation direction
[0198]	300	Imaging system
[0199]	302	Object to imaged
[0200]	304	Source
[0201]	306	Detector
[0202]	308	X-ray
[0203]	310	Structure within the object to be imaged
[0204]	311, 312	Rotation direction
[0205]	400	Flow chart 4D-CATH during the intervention
[0206]	402	Catheter placement
[0207]	403	Acquisition of update data
[0208]	404	Reconstruction of acquired data
[0209]	500	Flowchart of 4D-CATH
[0210]	501	Acquisition of high resolution CT scan

- [0211] 502 Performing intervention and acquiring update information
- [0212] 503 Incorporation of prior image
- [0213] 504 Reconstruction of image data
- [0214] 506 Collection of data
- [0215] 508 Reconstruction of soft tissue image; new prior
- [0216] 510 Continuous acquisition of low dose tomographic data during intervention
- [0217] 514 Situation X
- [0218] 512 Amount of tomographic data available
- [0219] 515 Situation X-t
- [0220] 516 Time axis
- [0221] 516.1, 516.2 Points in time
- [0222] 518.1, 518.2 Processing of data available
- [0223] 520 Reconstruction with further sparsifying function and/or prior image sparsifying
- [0224] 521 Reconstruction without further sparsifying function
- [0225] 524 Display of anatomy
- [0226] 526 Display of guide wires
- [0227] 528 Feedback for prior image
- [0228] 602 Prior image
- [0229] 604 Update scans
- [0230] 606 Projection at a solid angle for update scans
- [0231] 608 Projection at a solid angle for prior image
- [0232] 610 Incorporation of update scans in prior
- [0233] 700 PRIDICT reconstruction algorithm
- [0234] 702 Fully sampled set of projections
- [0235] 704 FDK
- [0236] 706 Prior image
- [0237] 708 Update image
- [0238] 710 Projector
- [0239] 712 Set of undersampled projections
- [0240] 714 Subtraction operation
- [0241] 716 FDK
- [0242] 718 Difference reconstruction
- [0243] 720 Summing operation
- [0244] 722 Subtraction operation
- [0245] 724 Image processing, mathematical operations
- [0246] 800 PRIDICT reconstruction algorithm including minimization
- [0247] 802 Fully sampled set of projections
- [0248] 804 FDK
- [0249] 806 Prior image
- [0250] 808 Update image
- [0251] 810 Projector
- [0252] 812 Set of undersampled projections
- [0253] 814 Subtraction operation
- [0254] 816 FDK
- [0255] 818 Difference reconstruction
- [0256] 820 Summing operation
- [0257] 822 Subtraction operation
- [0258] 824 Image to be minimized
- [0259] 826 Signum of image to be minimized
- [0260] 828 Subtraction operation
- [0261] 830 Comparator CS criterion
- [0262] 832 Image reconstruction loop
- [0263] 834 Minimization loop
- [0264] 900 PRIDICT reconstruction algorithm including sparsifying function
- [0265] 902 Fully sampled set of projections
- [0266] 904 FDK
- [0267] 906 Prior image
- [0268] 908 Update image
- [0269] 910 Projector
- [0270] 912 Set of undersampled projections
- [0271] 914 Subtraction operation
- [0272] 916 FDK
- [0273] 918 Difference reconstruction
- [0274] 920 Summing operation
- [0275] 922 Subtraction operation
- [0276] 924 Image to be minimized
- [0277] 926 Signum
- [0278] 928 Subtraction operation
- [0279] 930 Comparator CS criterion
- [0280] 932 Image reconstruction loop
- [0281] 934 Minimization loop
- [0282] 936 Further sparsifying function
- [0283] 1000 PRIDICT reconstruction algorithm including influence of reconstruction parameters
- [0284] 1002 Fully sampled set of projections
- [0285] 1004 FDK
- [0286] 1006 Prior image
- [0287] 1008 Update image
- [0288] 1010 Projector
- [0289] 1012 Set of undersampled projections
- [0290] 1014 Subtraction operation
- [0291] 1016 FDK
- [0292] 1018 Difference reconstruction
- [0293] 1020 Summing operation
- [0294] 1022 Subtraction operation
- [0295] 1024 Image processing, mathematical operation
- [0296] 1026 Sparsifying function
- [0297] 1028 Minimization
- [0298] 1030 Comparison of prior and update scans
- [0299] 1032, 1034 Influence on reconstruction algorithm
- [0300] 1040 Flow chart for adapting PRIDICT
- [0301] 1042 Continuous acquisition of undersampled data
- [0302] 1044 Calculation of amount of difference/movement
- [0303] 1048 Variation in reconstruction algorithm
- [0304] 1050 Providing a marker for the degree of completeness
- [0305] 1052 Input from radiologist
- [0306] 1046 Influencing scan parameters reconstruction parameters
- [0307] 1100 PRIDICT reconstruction algorithm including influence
- [0308] 1102 Fully sampled set of projections
- [0309] 1104 FDK
- [0310] 1106 Prior image
- [0311] 1108 Update image
- [0312] 1110 Projector
- [0313] 1112 Set of undersampled projections
- [0314] 1114 Subtraction operation
- [0315] 1116 FDK
- [0316] 1118 Difference reconstruction
- [0317] 1120 Calculation of modification
- [0318] 1122 Adaption of metric size
- [0319] 1124 Summing operation
- [0320] 1126 Subtraction operation
- [0321] 1128 Image to be minimized
- [0322] 1130 Signum
- [0323] 1132 Subtraction operation
- [0324] 1134 Comparator CS criterion
- [0325] 1136 Image reconstruction loop
- [0326] 1138 Minimization loop
- [0327] 1200 Guide wire

- [0328] 1300 Prior image
- [0329] 1302 Time frame reconstruction of update
- [0330] 1304 Time frame reconstruction of prior and update
- [0331] 1306 Time frame reconstruction of prior
- [0332] 1308 Heart to be fully imaged
- [0333] 1310 Cardiac phases
- [0334] 1312 Projections of prior
- [0335] 1314 Heart to be imaged through undersampled set
- [0336] 1316 Projection of undersampled set
- [0337] 1318 Cardiac phases
- [0338] 1320 Rotation
- [0339] 1322 Reconstructed prior
- [0340] 1324 Reconstructed update in cardiac phase 1
- [0341] 1326 Reconstructed update in cardiac phase 2
- [0342] 1328 Reconstructed update in cardiac phase 3
- [0343] 1330 Instrument
- [0344] 1332 Prior incorporated into each reconstruction of cardiac phases
- [0345] 1334 Reconstructed prior in cardiac phase 1
- [0346] 1336 Reconstructed prior in cardiac phase 2
- [0347] 1338 Reconstructed prior in cardiac phase 3
- [0348] 1340 Reconstructed update
- [0349] 1341 Instrument
- [0350] 1342 Prior for cardiac phases incorporated into reconstruction
- [0351] 1344 Reconstructed prior in cardiac phase 1
- [0352] 1346 Reconstructed prior in cardiac phase 2
- [0353] 1348 Reconstructed prior in cardiac phase 3
- [0354] 1350 Reconstructed update in cardiac phase 1
- [0355] 1352 Reconstructed update in cardiac phase 1
- [0356] 1354 Reconstructed update in cardiac phase 1
- [0357] 1356 Instrument
- [0358] 1358 Prior for cardiac phases incorporated into reconstruction of update for cardiac phases
- [0359] 1410 Full dose prior
- [0360] 1412 Low dose update
- [0361] 1414 4D data set of moving instruments
- [0362] 1416 4D intervention guidance in 3D road map
- [0363] 1418 Collected Low dose updates
- [0364] 1420 3D data set of vascular

1. An imaging method for radiologically guiding an instrument during medical interventions on an object, the method comprising:

- a) providing a first image of said object followed by
- b) providing updated images on-the-fly during the intervention to an operator by measuring an undersampled set of projections of said object and reconstructing said updated image based on changes between said first image or an update of said first image and said undersampled set of projections.

2. Method according to claim 1, wherein said imaging method is based on ionizing radiation and the undersampled set of projections is measured with a minimized radiation dose.

3. Method according to claim 1, wherein said first image, said update of said first image and said updated images comprise volumetric images of the object.

4. Method according to claim 1, wherein said first image comprises a high-resolution volumetric image of the object, which is acquired prior to the intervention, optionally based on a fully sampled set of projections, or which is provided from a database.

5. Method according to claim 1, wherein said update of said first image encompasses said first image and at least parts of one or more updated images reconstructed in previous runs during the intervention.

6. Method according to claim 1, wherein said update of said first image is a sliding prior defragmenting updated image of previous runs.

7. Method according to claim 1, wherein said undersampled sets of projections measured on-the-fly are stored for a delayed reconstruction of soft tissue contrast.

8. Method according to claim 1, wherein the amount of said changes is determined by forward projecting said first image or said update of the first image according to the projections comprised in the undersampled set of projections giving a set of equivalent projections based on the said first image or said update of the first image and subtracting equivalent projections.

9. Method according to claim 1, wherein the reconstruction includes an iterative minimization of a number of significant pixels optionally including at least one further sparsifying function.

10. Method according to claim 9, wherein the at least one further sparsifying function included in the reconstruction depends on the amount of data that has been acquired before the actual present.

11. Method according to claim 10, wherein updated images are reconstructed with no further sparsifying function and wherein the reconstruction of data stored from previous runs includes at least one sparsifying function.

12. Method according to claim 1, wherein the reconstruction is influenced by the amount of said changes between measurements.

13. Method according to claim 1, wherein the number of projections in the undersampled set of projections and the dose rate per projection are adapted according to the amount of changes between measurements and/or according to an input provided by the operator.

14. Method according to claim 1, wherein the reconstruction of said updated includes motion compensation.

15. Method according to claim 1, wherein periodic and/or non-periodic motion of the object is compensated for in the reconstruction of said updated images.

16. Method according to claim 15, wherein periodic motion compensation is performed through gating images into different phases of said periodic motion.

17. Method according to claim 15, wherein periodic motion compensation is performed through a transformation mapping said update images and/or said first image into one phase of said periodic motion.

18. Method according to claim 1, wherein said updated images are displayed on a screen allowing for different representation modes, which are chosen automatically or by said operator.

19. Method according to claim 1 for use in 4D radiological guidance.

20. Method according to claim 1 for use during interventions on the cardiovascular system, for use during catheter interventions and/or for use in the implantation of cardiac pacemakers.

21. Method according to claim 1 for use during interventions on tubular organ structures, preferably lungs or kidneys, for use in positioning of stents in vessels or bronchi or for use during bronchoscopy interventions.

22. Method according to claim **1** for use during interventions on the brain.

23. A system for radiologically guiding medical interventions on an object according to the method of claim **1**, the system comprising:

means to provide a first image of the object;

an imaging apparatus for measuring undersampled sets of projections;

processing means in communication with the imaging apparatus for providing updated images on-the-fly during the intervention by reconstructing said updated image based on changes between said first image or an update of said first image and said undersampled set of projections.

24. The system according to claim **23**, wherein the imaging apparatus is a tomographic system, such as a magnetic resonance imaging (MRI) scanner or a computed tomography (CT) scanner.

25. The system according to claim **24**, wherein the computed tomography scanner comprises at least one X-ray source and at least one detector.

26. The system according to claim **25**, wherein the X-ray sources differ in terms of the X-ray spectra or the detectors providing means of energy differentiation.

27. The system according to claim **23**, wherein imaging parameters of the imaging apparatus depend on the changes between said first image or an update of said first image and said undersampled set of projections.

28. The system according to claim **23** wherein the system includes means to provide an output to an operator, the output comprises at least one updated image, at least one image produced by delayed reconstruction of soft tissue contrast and/or an indicator for an accuracy of displayed images.

29. (canceled)

30. (canceled)

* * * * *

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摘要(译)

一种用于在对对象进行医学干预期间对器械进行放射学引导的成像方法，包括：提供所述对象的第一图像，然后在干预期间通过测量所述对象的欠采样的一组投影并重建，在运行中提供更新的图像。所述更新的图像基于所述第一图像之间的变化或所述第一图像的更新和所述欠采样的一组投影。本发明还涉及用于在物体上进行放射学引导医疗干预的方法和系统的特定用途，包括提供物体的第一图像；成像设备测量欠采样的投影组；与成像设备通信的处理装置通过基于所述第一图像或所述第一图像和所述欠采样投影组的更新之间的变化重建所述更新图像，在干预期间即时提供更新图像。

