(19) Europäisches Patentamt
European Patent Office
Office européen des brevets

(11) EP 2 146 321 A1

(12) **EUROPEAN PATENT APPLICATION**

(43) Date of publication:
20.01.2010 Bulletin 2010/03

(51) Int Cl.:
G06T 11/00 *(2006.01)*

(21) Application number: **08160359.9**

(22) Date of filing: **14.07.2008**

(84) Designated Contracting States:
AT BE BG CH CY CZ DE DK EE ES FI FR GB GR HR HU IE IS IT LI LT LU LV MC MT NL NO PL PT RO SE SI SK TR
Designated Extension States:
AL BA MK RS

(71) Applicant: **Cefla S.C.**
40026 Imola (IT)

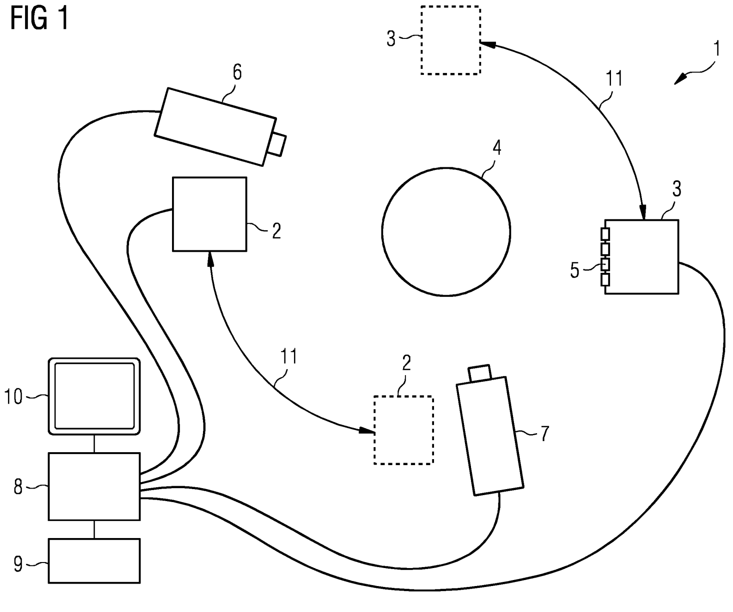
(72) Inventors:
- **Borghese, Nunzio Alberto**
20129 Milano (IT)
- **Rotondo, Giuseppe**
20090 Pantigliate - Milano (IT)
- **Nanni, Eros**
40024 Castel S. Pietro (IT)

(74) Representative: **Herrmann, Franz**
Dendorfer & Herrmann
Patentanwälte Partnerschaft
Bayerstrasse 3
80335 München (DE)

(54) Dynamic error correction in radiographic imaging

(57) A method for radiographic imaging is described. Radiographic raw images of an object (4) are generated by an imaging system comprising an X-ray source (2) and an X-ray detector (3) and moving along a given trajectory (11) during an acquisition process of the radiographic images. During the generation of the radiographic raw images displacement data describing the displacement of the moving object (4) are generated by means of a sensor arrangement (6, 7) and a control unit (8) connected with the sensor arrangement (6, 7). Finally views of the object (4) are produced by image processing of the radiographic raw images in the control unit (8) wherein the displacement data are used for compensating the motion of the object (4). For the compensation of the motion, the object (4) is treated as stationary and the actual trajectory (11) is replaced by a virtual trajectory resulting in a spatial relation between the stationary object (4) and the imaging system on the virtual trajectory that corresponds to the spatial relation between the displaced object (4) and the imaging system on the actual trajectory (11).

FIG 1



Printed by Jouve, 75001 PARIS (FR)

Description

[0001] The invention relates to a method for radiographic imaging, the method comprising the steps of:

- generating radiographic raw images of an object by means of an imaging system comprising an X-ray source and an X-ray detector and moving along a trajectory during an acquisition process of the radiographic raw images;
- generating displacement data describing the displacement of the moving object during the generation of the radiographic raw images by means of a sensor arrangement and by means of a control unit connected with the sensor arrangement;
- producing a view of the object by image processing of the radiographic raw images in the control unit wherein the displacement data are used for compensating the motion of the object.

[0002] The invention further relates to an apparatus for implementing the method.

[0003] Such a method and such an apparatus is known from US 71 87 749 B2. The known apparatus is provided for dental panoramic and tomographic imaging. According to the known method and the known apparatus the motion of a patient is detected during the imaging process by accelerometers attached to a positioning device and the radiographic raw images are shifted to correct for the motion of the patient.

[0004] A disadvantage of the known method and the known apparatus is that the shifting of the radiographic raw images is only suitable for compensating translational movements. The lack of compensation for arbitrary movement of the patient during the imaging process may result in blurring of the views obtained by the image processing.

[0005] In the field of tomography, blurring by motion is a common problem. For limiting the blurring effects patient support systems are usually provided for the purpose of holding body parts of the patient as still as possible with respect to the X-ray apparatus. For instance, in dental panoramic X-ray systems, which are also arranged for dental tomography, a bite may be provided, on which the patient can bite during the entire acquisition to limit the motion of the patient. In alternative arrangements, the patient lies on a bed and the body parts are kept still with the help of radio transparent head cushions and belts. However, even these tools are not sufficient for suppressing the motion of the patient completely.

[0006] EP 1 815 794 A1 discloses a positioning method for dental radiology and linear tomography. For positioning the patient with respect to an X-ray imaging system a set of video cameras is used to put the patient in a given desired position for the examination.

[0007] KAK & SLANEY, Principle of Computerized Tomographic Imaging, SIAM Society, 2001 is a text book containing details on various reconstruction methods used in computer-tomography, in particular frequency based reconstruction methods and various algebraic methods, for instance the so called Algebraic Reconstruction Techniques (= ART) and its derivatives, especially the Simultaneous Iterative Reconstruction Technique (=SIRT) and Simultaneous Algebraic Reconstruction Technique (=SART).

[0008] A general review on computer-tomography can be found in KALEDENDER, W. A.: X-ray computed tomography in: Phys. Med. Biol. 51 (2006) R29-R43.

[0009] MICHEL, D.; FRÉDÉRIC; HIROYUKI, K.: Two-dimensional rebinning of helical cone-beam computerized tomography data using John's equation, Inverse Problems, Volume 19, Issue 6, pp. S41-S54, 2003 describes a rebinning method for compensating the deviation of an x-ray imaging system moving around an object on a helical path from a two-dimensional slice on which a crossectional view of the object is generated.

[0010] DOBBINS III, J.T. and GODFREY, D.J.: Digital X-ray tomosynthesis: current state of the art and clinical potential, Phys. Med. Biol. 48 (2003) R65-R106 describes various reconstruction methods used in tomosynthesis.

[0011] SHEPP, L. A. and VARDI Y.: Maximum Likelihood Reconstruction for Emission Tomography, IEEE Trans. Med. Im., Vol. 1, 113-122, 1982 describes the application of a maximum likelihood method in positron emission tomography.

[0012] GB 23 84 155 A describes a video system which is used to align tomographic images of a patient to a radiotherapy system so that the tomographic images can be used for controlling the beam of the radio-therapy-system.

[0013] US 2006/0149134 A1 discloses a visual assisted guidance of an endoscope for bronchoscopy.

[0014] The use of a maximum likelihood approach for the deconvolution of images taken by a CCD-camera is discussed in BEN-VENUTO, F. and LA CAMERA, A. et al.: Study of an iterative method for the reconstruction of images corrupted by Poisson and Gauss noise, Inverse Problems 24, 2008.

[0015] An overview on biometric systems is given in JAIN, A.K. and ROSS, A.: Multibiometric Systems, Comm. ACM, Special Issue on Multimodal Interfaces, vol. 47, no. 1, pp. 34-40, 2004.

[0016] WISKOTT, J. et al.: Face recognition by elastic bunch graph matching. In L.C. Jain et al., editor: Intelligent biometric techniques in fingerprints and face recognition, pp. 355-396. CRC Press, 1999 describe methods for face recognition.

[0017] Further information of face recognition can also be found in FERIS, R.S. et al.: Hierarchical wavelet networks for facial feature localization, IEEE International Conference on Automatic Face and Gesture Recognition, 2002, pp.

118-123.

[0018] MIDDENDORFF, C.; BOWEYER, K.; YAN, P.: Multi-Modal Biometrics Involving the Human Ear, Proc. IEEE Conf. CVPR 07, 2007 describes methods for detecting the human ear.

[0019] MA, Y. et al.: Robust precise eye location under probabilistic framework, IEEE International Conference on Automatic Face and Gesture Recognition, 2004, pp. 339-344 discloses methods for the detection of eyes.

[0020] DARGHAM, J. and CHEKIMA, A.: Lips Detection in the Normalised RGB Colour Scheme Proc. 2nd Conf. on Information and Communication technologies, ICCTA, 2006, pp. 1546-1551, 2006 contain methods for detecting the lips of a person.

[0021] FROSIO, I. and BORGHESE, N.A.: Real-time accurate circle fitting with occlusions, Vol. 41, pp. 101-114, 2008 describes a method for an accurate location of circles inside images in the presence of a partial occlusion.

[0022] Proceeding from this related art, the present invention seeks to provide a method and an apparatus for digital radiography with improved motion compensation.

[0023] This object is achieved by a method having the features of the independent claim. Advantageous embodiments and refinements are specified in claims dependent thereon.

[0024] In the image processing an operator is used, which relates a plurality of radiographic raw images with the view of the object without transforming the radiographic images individually. The displacement of the object during the acquisition process is then taken into account by modifying the operator in dependency of the displacement data. By the method, the radiographic raw images can be used in the imaging process without any modifications which might induce errors in the reconstruction process. Furthermore, even complex movement can be taken into account by modifying the operator relating the plurality of radiographic raw images with the view, so that blurring can be effectively suppressed.

[0025] In the imaging process the displacement of the object during the acquisition process is taken into account by considering the object as stationary and replacing the actual trajectory by a virtual trajectory resulting in a spatial relation between the stationary object and the imaging system on the virtual trajectory that corresponds to the spatial relation between the displaced object and the imaging system on the actual trajectory. By the method, the radiographic raw images can be used in the imaging process with the appropriate spatial information concerning the relative location and orientation of the object with respect to the imaging system. Thus, the reconstruction process can be performed while taking into account the known relative position and orientation of the object with respect to the imaging system. By the method, even a complex movement can be appropriately treated, so that blurring can be effectively suppressed.

[0026] Preferably, the displacement data are associated with the individual radiographic raw images and describe the displacement of the object with respect to a previous position of the object at the exposure time of a previous radiographic raw image. Thus, each radiographic raw image can be used for the reconstruction process.

[0027] The control unit can be arranged for generating displacement data describing the rotational and translational displacement of the object during the acquisition process. Thus, the movement of the object can be completely tracked.

[0028] The displacement data can be contained in an affine displacement matrix that is determined by the control unit. In this case the virtual trajectory is created by applying the inverse of the affine displacement matrix to the actual trajectory.

[0029] The displacement data are preferably derived from surface images of the object. The surface images can be acquired by a two-dimensional or three-dimensional vision capture system and the information on the position and orientation of the object are derived from the surface images by the control unit. The surface images acquired by the vision capture system can be processed by the control unit using image registration and image subtraction, pattern analysis, template matching or curve fitting for deriving the displacement.

[0030] The displacement data can also be produced by means of three-dimensional scanners and the control unit can utilize three-dimensional surface matching or three-dimensional real time cross-correlation analysis for determining the displacement of the object.

[0031] The acquisition of spatial information data can be made more reliable, if markers placed on the object are detected by the sensor arrangement.

[0032] The data supplied by the sensor arrangement can be processed by the control unit with statistical techniques so that the most likely spatial relation between object and imaging system is determined. This method is particularly useful, if the data supplied by the sensor arrangement is affected by noise or if patterns must be recognized for the retrieval of the displacement data.

[0033] The sensor arrangement may comprise a sensor capable of detecting the actual acceleration or velocity of the object since acceleration sensors or velocity sensors are generally reliable devices for detecting the motion of an object. In this case the control unit determines the resulting displacement of the object by integrating the signal supplied by the acceleration or velocity sensor.

[0034] For improving the accuracy of the estimation of the motion of the object a time filtering is performed on the sequence of the displacement data based on extended Kalman filtering.

[0035] For adjusting the spatial position of the object with respect to the imaging system an object support system is utilized to position the object in a predefined position with respect to the imaging system before starting the X-ray exposure.

[0036] In a preferred embodiment the imaging process is a reconstruction process resulting in tomographic images.

[0037] The reconstruction process can be based on an algebraic method for solving a system of equations. These system of equations connects the absorption coefficients associated with volume elements of the object with projections values of radiographic images taken at various positions along the trajectory. The equations system is formed by means of weighted sums of the absorption elements along radiation paths associated with the projection values of the detector elements of the detector. The weighting coefficients depend on the length of a segment of a given radiation path lying within a given volume element. Therefore, the weighting coefficients depend on the direction of the radiation path of the radiation beams. For taking into account the virtual movement of the imaging system along the virtual trajectory the weighting coefficients are modified in dependency of the displacement data resulting in an image reconstruction with suppressed blurring effects.

[0038] In another preferred embodiment, a frequency based reconstruction process is used in which the absorption coefficient associated with a given volume element are obtained by applying a backprojection on the filtered radiographic raw images taken from various positions along the trajectory, wherein the backprojection is performed along the radiation paths associated with the imaging system on the virtual trajectory. This kind of imaging processing can be applied even if the virtual trajectory deviates by a large amount from the actual trajectory.

[0039] If the deviation of the virtual trajectory from the actual trajectory is small a frequency based reconstruction process can be used, in which the absorption coefficient associated with a given volume element are obtained by applying a rebinning process on the radiographic images. By the rebinning process the projection data of a two-dimensional slice of the object are estimated on the basis of the displacement data. After the rebinning process has been performed a two-dimensional reconstruction process can be applied to the projection data for reconstructing the required two-dimensional view of the object.

[0040] The object is preferably a body of a patient and the imaging system is preferably arranged for cone beam computer tomography, tomosynthesis, dental tomography, dental panoramic radiography, cephalography or medical scanography.

[0041] Further advantages and properties of the present invention are disclosed in the following description, in which exemplary embodiments of the present invention are explained in detail based on the drawings:

Figure 1	illustrates an arrangement of devices used in digital tomographic imaging;
Figure 2	shows the relative position of a reference system and an imaging system with respect to an object which has to be examined;
Figure 3	illustrates the spatial relationship of the reference system and the imaging system with respect to the object after the object has moved; and
Figure 4	demonstrates the effect of various transformations used to describe and to compensate for the movement of the object.

[0042] The method described here is used in a medical apparatus 1, for medical X-ray imaging, which can be used for tomographic imaging. The apparatus 1 can also be used for panoramic imaging. Such an apparatus has an X-ray emitter 2 and a X-ray detector 3 opposite one to the other, and an object 4 to be surveyed has to be placed between emitter 2 and detector 3. In the case of a medical apparatus 1 the object is a body part of a patient. The emitter 2 is generally an X-ray tube whereas the detector 3 is a digital detector comprising a plurality of detector elements 5. Correspondingly, the radiographic image is composed of a number of picture elements associates with the detector elements 5. These picture elements are also called pixels.

[0043] For tracking the motion of the object 4 during the acquisition of the radiographic images, the apparatus 1 is provided with video cameras 6 and 7.

[0044] The emitter 2, the detector 3 and the video cameras 6 and 7 are connected to a control unit 8 which is provided with an input device 9 such as a keyboard, a mouse or a similar device. The control unit 8 is further connected to a display 10 on which views of the object 4 can be displayed for the operator. In the case of the apparatus 1 these views can be panoramic images of the object 4, for instance panoramic images of the dental arc of a patient, or tomographic images of the object 4, for example a two- or three-dimensional tomographic image of a single tooth.

[0045] For tomographic imaging, the emitter 2 and the detector 3 are moved along a given trajectory 11. Along the trajectory, N radiographic images are acquired by the detector 3 at different positions resulting in N radiographic images taken under various projection directions.

[0046] The tomographic image of the object 4 can be reconstructed by applying a reconstruction procedure on the N radiographic images. If the trajectory 11 of the emitter 2 and the detector 3 is situated in a plane, the tomographic image corresponds to a two-dimensional cross-section through the object 4 along the plane. However, if the trajectory 11 is situated in a three-dimensional space the tomography algorithms generally transform a set of N two-dimensional X-ray images into a three-dimensional volumetric image.

[0047] It should be emphasized that the two-dimensional radiographic images are usually acquired sequentially, one after the other, therefore in different instants of time.

[0048] Figure 2 shows a cross-section through the object 4. The cross-section extends in an XY-plane of a reference system 12. The projection value at the position p(x,y) of the radiographic image taken by the detector 3 represent the attenuation of an X-ray 13 through the object 4. In particular, the projection value taken by the detector 3 can be represented as the line integral of the attenuation coefficient of the material of the object 4. In particular the following relationship holds:

$$\ln\frac{N_e}{N_m} = \int_{ray} \mu(X,Y,Z)ds \qquad (1)$$

where $\mu(X,Y,Z)$ is the linear absorption coefficient measured in the point P(X,Y,Z) within the object 4 and the ray is the straight line followed by the X-ray 13. N_e is the number of photons emitted by the emitter 2 and N_m the number of photons measured on the detector 3.

[0049] The overall X-ray attenuation along the X-ray 13 is measured by an pixel value in position p(x, y). In praxis the overall X-ray attenuation is represented as a gray level g(.). The gray level g(p(x,y)) is equal to:

$$g(p(x,y)) = N_e e^{-\int_{ray} \mu(X,Y,Z)ds} \qquad (2)$$

[0050] By taking the logarithm of the pixel value g(.) the following linear relationship between the negative logarithm of the pixel value and the absorption coefficients is obtained:

$$-\ln(g(p(x,y))) = -\ln(N_e) + \int_{ray} \mu(X,Y,Z)ds \qquad (3)$$

where the transfer function of the detector element 5 associated with the pixel p(x, y) is omitted for sake of clarity. If the number of photons emitted in all directions is the same, equation (3) can be simplified to:

$$-\ln(g(p(x,y))) = \int_{ray} \mu(X,Y,Z)ds \qquad (4)$$

[0051] In tomography, the position and orientation of the X-ray emitter 2 and of the detector 3 is supposed to be precisely known with respect to the reference system 12 for each acquired X-ray image. The spatial relationship of the emitter 2 and the detector 3 with respect to the reference system is called nominal geometry.

[0052] With respect to the reference system 12, that is fixed with respect to the apparatus 1, a tomographic volume 15 can be defined as a set of adjacent voxels 16, of given size, whose position is precisely known in the reference system 12: $V_j(\mathbf{P}_{cj}(X_j,Y_j,Z_j))$, where $\mathbf{P}_{cj}$ is the voxel center. The field of the absorption coefficients can be described by assigning an absorption coefficient to each voxel 16. For instance, the mean absorption coefficient $a(V_j(X_j,Y_j,2_j))$ inside a voxel V_j can be attributed to the voxel V_j:

$$a_j(V_j) = \frac{1}{V} \int_{\vec{x} \in V_j} a(\vec{x})\,dV \qquad (5)$$

[0053] An alternative representation can be obtained through basis functions, which allow avoiding discontinuities at the voxels borders. Such an approach is disclosed in SCHWEIGER, M. and ARRIDGE, S.R., Image reconstruction in optical tomography, using local basis functions, J. Electronic Imaging, Volume 12, Issue 4, pp. 583-593, 2003.

[0054] The projection equation (4) can also be discretized and written as:

$$-\ln g(p(x,y)) = \sum_{j \in ray} w_j a_j\left(V_j\right) \qquad (6)$$

[0055] The weights, $\{w_j\}$, in equation (6) express the partial volume effect due to the fact that the intersection of the X-ray 13 with the j-th voxel 16 has length w_j. We remark here, that given a certain nominal geometry the weights are uniquely determined as they depend only on the acquisition geometry.

[0056] Therefore, equation (6) can also be written for the i-th pixel in the n-th radiographic image as:

$$-\ln g_i(\vec{s}_n)) = \sum_{j \in ray_i} w_j(\vec{s}_n) a_j\left(V_j\right) \qquad (7)$$

where $\vec{s}_n$ is the vector between the emitter 2 and the i-th pixel 5 of detector 3 at step n along the trajectory 11.

[0057] A set of weights is associated to each single radiographic image. These weights can be computed, for instance, from the intersection of a voxel 16 with the radiation path 14 connecting the center of each detector element 5 with the X-ray emitter 2.

[0058] The set of weights and the tomographic volume 15 are therefore uniquely determined once the nominal geometry has been defined.

[0059] Most of the techniques used for tomographic reconstruction are based on the hypothesis that a set of radiographic images are acquired in a known and equally spaced position on an arch of at least 180 degrees. Under this hypothesis, techniques in the frequency domain, based on the Radon transform, are used and produce a tomographic reconstruction in a reduced amount of time. This approach leads to the methods which can be applied to the case of fan shaped X-ray beams and cone shaped X-ray beams.

[0060] Another group of methods used to reconstruct tomographic images, are iterative techniques. These can be further subdivided into two large families: algebraic and statistically based techniques. Both approaches are based on equation (7).

[0061] Algebraic techniques, in particular the ART methods and its derivatives SART and SIRT produce a tomographic image by solving a very large linear system of *N*M* equations (7), one for each of the *M* pixels constituting the *N* radiographic images acquired. The unknowns of the equation system are the absorption coefficients $\{a_j\}$.

[0062] In statistical based techniques a tomographic image is obtained as the one which maximizes the likelihood of the data, possibly adding constraints to the solutionThis is an approach which was started in positron emission tomography (= PET) imaging and astrophysics and which is also applicable to the field of X-ray radiography and tomography.

[0063] The last group of tomographic methods is represented by tomosynthesis. In this family of methods the reconstruction is obtained as a set of parallel planes or slices. The algorithm is quite simple as it is based on shifting and eventually scaling, by an adequate amount the images, and adding them together.

[0064] The underlying hypothesis of all these methods is that the object 4 is keeping still during the entire acquisition that is from the acquisition of the first radiographic image to the acquisition of the N-th one. Therefore the object 4 is supposed to have the same eventually unknown position and orientation inside the tomographic volume 15. In praxis, however, the object 4 is rarely keeping still. In particular, a living organism hardly keeps still for a while. The motion of the object 4 also affects the quality of the view, which is generated from the radiographic images.

[0065] Figure 3 depicts the situation that the object has moved with respect to the situation shown in Figure 2. When the object 4 is moving, the object region crossed by a certain X-ray 13 denoted by r in Figure 3 and projected onto a point p(x,y) with the nominal geometry, is crossed by a different X-ray 13 denoted s and projected onto a different point p'(x,y) of the radiographic image.

[0066] The integral absorption measured at site p(x,y) is therefore attributed to a wrong voxel due to the motion of the object 4, and this distorts the reconstruction and produces what is called blurring by motion in the reconstructed tomographic image.

[0067] It should be noted that a typical voxel size of 100 μm is required to see the anatomical details in a dental radiography; therefore blurring due to motion should stay inside this range. Therefore any blur compensation mechanism should achieve such accuracy.

[0068] Let us suppose that the object 4 had moved rigidly in the time interval between two radiographic images. The motion can be described by an affine matrix **A**, which contains the translation vector and the rotation matrix as depicted in Figure 4. All the points **{P}** of the object 4 are moved to points given by **AP,** in homogeneous coordinates. The voxel centers $\mathbf{P_{cj}}$ are moved accordingly by $\mathbf{AP_{cj}}$**.** Details on the construction of such affine matrices can be found in HARTLEY, R. and ZISSERMAN, A.: Multiple view geometry in Computer Vision, 2nd edition, Cambridge University Press, 2004.

[0069] In particular, n affine matrices $\mathbf{A_n}$ which corrects for the motion of the object 4 are uniquely defined for each radiographic image.

[0070] According to equation (7) the effect of the displacement of the object 4 can then be expressed as:

$$-\ln g_i(\vec{s}_n)) = \sum_{j \in ray_i} w_j(\vec{s}_n) a'_j (V_j) \qquad (8)$$

with

$$a'_j(V_j) = \frac{1}{V} \int_{\vec{x} \in V_j} a(A_n^{-1}\vec{x})dV \qquad (9)$$

[0071] Therefore, the number of unknowns is largely increased since the coefficients a'_j are associated with a region which is different from the region associated with the coefficient a_j.

[0072] With the knowledge of the matrix **A**, the ray s passing through the voxel center $\mathbf{P_{cj}}$ and the emitter position $\mathbf{P_e}$ and the intersection point p'(x,y) of the X-ray s with the image plane of the detector 3 can be determined after the object 4 has moved. The point p'(x,y) depends both on the motion of the object 4 and on the geometry of the acquisition system, namely the relative distance between the object 4, the emitter 2 and the image plane of the detector 3.

[0073] In practical applications, for each X-ray 13 through the image plane of the detector 3 and the center of the emitter 2, the voxels 10 crossed after the motion of the object 4 can be determined along with the new length w_j of the radiation path covered inside the j-th voxel.

[0074] This allows rewriting the set of projection equations (7), where the set of voxels 16 crossed by the ray has been redetermined along with their associated weights. Equation (7) then reads:

$$-\ln g_i(\vec{s}\,'_n)) = -\ln g_i(A_n^{-1}\vec{s}_n) = \sum_{j \in ray_i} w_j(A_n^{-1}\vec{s}_n) a_j (V_j) \qquad (10)$$

[0075] In equation (10) the affine matrix $\mathbf{A_n}$ has been applied to the reference system 12 as well resulting in a virtual trajectory that originates from the actual trajectory in the original reference system by applying the inverse $\mathbf{A_n^{-1}}$ of the affine matrix $\mathbf{A_n}$ to the actual trajectory. This is nothing else than treating the object as stationary and replacing the actual trajectory by a virtual trajectory which results from applying the inverse $\mathbf{A_n^{-1}}$ of the affine matrixes $\mathbf{A_n}$ to the positions of the actual trajectory.

[0076] The solution of equation (10) then proceeds using standard techniques associated to algebraic, statistical or tomosynthesis methods.

[0077] It should be noted that the apparatus 1 can also acquire a sequence of radiographical images under a limited angle to produce a three-dimensional local tomographic image. Such a mode of operation is also called limited angle tomography. Alternatively it can be used to create a panoramic radiography.

[0078] Iterative methods and tomosynthesis do not pose any constraint on the geometry of the trajectory used for the acquisition of the radiographic images and are therefore suitable methods for the compensation of the motion of the object 4.

[0079] However, the basic approach of the method described above can also be applied to frequency spaced methods, which generally require a particular spatial arrangement of the radiographic images, for instance the radiographic images to be equally spaced along the trajectory.

[0080] In the context of the frequency based methods the measured radiographic images are also called projection images. According to one method called filtered backprojection, these projection images can be filtered by convolution with the kernel of the Hilbert transform and are then backprojected along the x-rays 13 of the radiation fan. If the moved object 4 is considered as stationary with respect to the reference system 12 whereas the actual trajectory 11 is replaced by a virtual trajectory resulting in a spatial relation between the stationary object 4 and the imaging system on the virtual trajectory that is the same as the spatial relation between the displaced object 4 and the imaging system on the actual trajectory 11, the backprojection can be performed along the x-rays 13 of the radiation fan originating from the virtual position of the emitter 2. This approach can be used independent from the extent of the motion of the object 4.

[0081] If the motion of the object 4 is sufficiently small the so called rebinning methods can also be used. These methods are generally used for tomographic imaging in which the imaging system is moving around the object on a helical path. In these cases, the position of the emitter 2 of the radiation fan moving around the object deviates from the image plane in which the view of the object is generated and a correction for the deviation of the actual trajectory from the image plane is determined by the rebinning methods. The deviation of the virtual trajectory from the actual preset trajectory 11 can then be treated similar to the deviation of the helical path from the image plane.

[0082] For tracking the motion of the object 4, the apparatus 1 is endowed with a pair of video cameras 6 and 7 positioned at the side of the trajectory 11, as depicted in Figure 1. The video cameras 6 and 7 are positioned at an angle slightly larger than ninety degrees, observing the object 4. The cameras 6 and 7 are supposed to be registered to the apparatus 1 so that the position of the detector of the video cameras 6 and 7 is available in the same reference system 12 in which, the trajectory 11 of the X-ray emitter 2 and of the detector 3 is given.

[0083] Image processing techniques are used to detect the motion of the object 4. In case of dental tomography the object 4 is the head of the patient and the motion of the head is measured on the image plane of each camera 6 or 7.

[0084] Among the image processing techniques, image subtraction is implemented in a preferred embodiment to get the rough estimate of the displacement of the object 4 with respect to each radiographic image. For image subtraction an image of the background without the object is subtracted from an image of the background with the object 4.

[0085] This rough estimate can then be refined by using image processing techniques tailored to identify specific head anatomical features. These techniques allow to better identify the head displacement on the images of the video cameras, in particular the two-dimensional displacement.

[0086] In particular flexible template matching can be used to accurately identify the ear, surveyed by a lateral camera 12, and track the motion of the head in the sagittal plane.

[0087] Template matching and curve fitting can be used to accurately track the eyes pupils. The pupils position can be used to initialize a curve which can be used to fit the eye profile. By the eye profiles the motion of the head in a coronal plan can be tracked.

[0088] A similar approach can be used to identify the lips profile.

[0089] From the multiple displacement estimates obtained, for instance, with the above mentioned techniques, the three-dimensional motion of the object 4 can be derived by standard photogrammetric techniques. It should be noticed that in most cases only the translational component of the motion will be detected, as for very small movement as those foreseen when the object 4 is requested to stay still, the rotation and translational component can be lumped together.

[0090] During this process, the different sensitivity to the different degrees of freedom offered by each feature should be taken into account. For instance, the eye profile is quite accurate to identify the vertical position, while it is may not be so precise in the location of the horizontal position. This allows obtaining the maximum accuracy. The two-dimensional measurements got from the different features can therefore be pulled together and weighted to derive the head displacement onto the image.

[0091] In a modified embodiment only one video camera 6 or 7 is provided. In this case, motion in only one plane can be tracked. This can be adopted in a machine, whose head resting frame would constraint head motion to a plane.

[0092] In a different embodiment, the apparatus can be equipped with one or more three-dimensional cameras. In this case, three-dimensional surface matching techniques or three-dimensional cross-correlation can be used to detect head motion. These techniques could be used in addition or not to feature detection.

[0093] In an alternative embodiment, a set of radiolucent markers are attached to the head of the patient. In this case the features extracted from the images, are the markers themselves, and the position of the features is coincident with the center of the marker. High accuracy statistical techniques can be used for this scope as disclosed in FROSIO and BORGHESE. In this embodiment, markers can be placed for instance on an elastic band which can be easily wore by the patient.

[0094] Other types of sensors can be used with or without the video cameras 6 and 7 to track the displacement of the head.

[0095] In an alternative embodiment, the radiolucent markers can be substituted by active radiolucent markers, which actively signal their position in three-dimensional space. This can be done through radio-frequency transmission, for instance by RF-ID tags.

[0096] Motion of the object can also be determined by one of the many motion capture technologies available. For instance but not exclusively, marker-based motion capture, active markers motion capture, accelerometers and gyroscope-based motion capture, computer vision based techniques can be used. Other methods uses techniques like real-time three-dimensional cross-correlation of three-dimensional images acquired by three-dimensional cameras.

[0097] When multiple rigid objects have to be considered, for instance the skull and the mandible, which move differently, the above procedure can be generalized. The motion of the multiple objects can be tracked separately, with the constraint of congruency of motion. In this case a set of matrixes, $\{\mathbf{A_B}\}$ is obtained, one for each object B, which is used to correct the relative position of that particular object, with respect to the apparatus reference frame.

[0098] A further improvement of the method can be obtained by considering each matrix $\mathbf{A_m}$ as an additional unknown inside the computing machinery used to produce the output image. This machinery usually boils down into optimization procedures. Therefore, in this case, the motion tracking apparatus, if present, produces a reliable initialization of the patient position in each instant of time.

[0099] A time filtering on the sequence of the estimated displacements can be inserted, based on extended Kalman filtering, to improve the accuracy of the estimate of the patient motion.

[0100] This method described herein can be applied to all imaging techniques in which the final view is obtained from a set of raw images acquired sequentially in time and for which the absence of motion of the object cannot be guaranteed. Domains of application can be panoramic imaging, fan beam or cone beam CT or computer assisted tomography, digital panoramic radiography and digital cephalography and all those fields where the output image is obtained starting from a set of X-ray images and or image sensor data acquired sequentially.

[0101] The method can be applied to any imaging system of an object which produces an output view from a set of raw images, each of which needs a geometrical correction, which can be computed through motion capture of the object.

[0102] From an abstract point of view, the computational operations needed for the image processing can be summarized into a single operator which relates the raw images with the final view:

$$h_j = F_j(\vec{s}_1, \dots, \vec{s}_N)[(g_i(\vec{s}_1), \dots, g_i(\vec{s}_N))_{i \in \det ector}] \qquad (11)$$

where h_j is the j-th image value of the final view, g_i is the value of the i-th pixel of the raw images, $\vec{s}_n$ is the vector between the emitter and the i-th pixel of the detector at step n along the trajectory and $F_j(\vec{s}_1,\dots,\vec{s}_N)$ is the operator relating the j-th image value of the final image with the pixel values of the raw images at the positions represented by $\vec{s}_1$ up to $\vec{s}_N$. In the case of a tomographic reconstruction with algebraic methods, h_j is the absorption coefficient associated with the voxels j and equation (11) is nothing else than the equation (7) solved for the absorption coefficients a_j.

[0103] In case of a displacement of the object at step n along the trajectory the disturbance can be expressed as:

$$h'_j = F_j(\dots, \vec{s}_n, \dots)[(\dots, g'_i(\vec{s}_n), \dots)_{i \in \det ector}] \qquad (12)$$

which can be compensated by replacing the actual trajectory by a virtual trajectory:

$$h_j = F_j(\dots, A^{-1}\vec{s}_n, \dots)[(\dots, g'_i(A^{-1}\vec{s}_n), \dots)_{i \in \det ector}] \qquad (13)$$

[0104] Thus, the disturbance is compensated by the modification of the operator F_j.

[0105] It should be noted that F_j needs not to be an operator used for the reconstruction of tomographic images but may also be an operator used for tomosynthesis, dental panoramic radiography, cephalography or medical scanography.

[0106] The method and the apparatus described herein provide a radiographic apparatus endowed with a system capable of tracking the motion of the object 4 in real-time and a method to derive from the motion information an efficient correction for each acquired radiographic image.

[0107] The great advantage obtained by this method is evident as it would allow to stabilize the object 4 of the tomography and therefore to potentially increase the resolution and the clarity of the tomographic images.

[0108] Throughout the description and claims of this specification, the singular encompasses the plural unless the context otherwise requires. In particular, where the indefinite article is used, the specification is to be understood as contemplating plurality as well as singularity, unless the context requires otherwise.

[0109] Features, integers, characteristics, compounds or groups described in conjunction with a particular aspect, embodiment or example of the invention are to be understood to be applicable to any other aspect, embodiment or example described herein unless incompatible therewith.

Claims

1. A method for radiographic imaging, the method comprising the steps of:

- generating radiographic raw images of an object (4) by means of an imaging system comprising an X-ray source (2) and an X-ray detector (3) and moving along a trajectory (11) during an acquisition process of the radiographic raw images;
- generating displacement data describing the displacement of the moving object (4) during the generation of the radiographic raw images by means of a sensor arrangement (6, 7) and by means of a control unit (8) connected with the sensor arrangement (6, 7);
- producing a view of the object (4) by image processing of the radiographic raw images in the control unit wherein the displacement data are used for compensating the motion of the object (4),

characterized in that
in the image processing an operator is used, which relates a plurality of radiographic raw images with the view of the object (4) without transforming the radiographic images individually, and that the displacement of the object (4) during the acquisition process is taken into account by modifying the operator in dependency of the displacement data.

2. The method according to Claim 1,wherein in the image processing the displacement of the object (4) during the acquisition process is taken into account by treating the object (4) as stationary and replacing the actual trajectory (11) by a virtual trajectory resulting in a spatial relation between the stationary object (4) and the imaging system on the virtual trajectory that corresponds to the spatial relation between the displaced object (4) and the imaging system on the actual trajectory (11).

3. The method according to Claim 1 or 2,
wherein the displacement data are associated with the individual radiographic raw images and describe the displacement of the object (4) with respect to a previous position of the object (4) at the exposure time of a previous radiographic raw image.

4. The method according to Claim 1 or 3,
wherein the control unit (8) is generating displacement data describing the rotational and translational displacement of the object (4) during the acquisition process.

5. The method according to any one of Claims 1 to 4,
wherein an affine displacement matrix is determined by the control unit (8) as displacement data and that the virtual trajectory is created by applying the inverse of the affine displacement matrix to the actual trajectory (11).

6. The method according to any one of Claims 1 to 5,
wherein the sensor arrangement comprises a two-dimensional or three-dimensional vision capture system (6, 7) which is used for acquiring surface images of the object (4) and wherein the displacement data are derived from the surface images.

7. The apparatus according to Claim 6,
wherein the control unit (8) processes the surface images by image registration and image subtraction, pattern analysis, template matching or curve fitting.

8. The apparatus according to any one of Claims 1 to 7, wherein the sensor arrangement comprises three-dimensional scanners and wherein the control unit (8) utilizes three-dimensional surface matching or three-dimensional real time cross-correlation analysis for determining the displacement data.

9. The apparatus according to any one of Claims 1 to 8,
wherein markers placed on the object (4) are detected by the sensor arrangement.

10. The apparatus according to any one of Claims 1 to 9,
wherein the data supplied by the sensor arrangement are processed by the control unit (8) with statistical techniques for determining the displacement data.

11. The apparatus according to any one of Claims 1 to 10,
wherein the actual acceleration or velocity of the object (4) is detected by the sensor arrangement and wherein the control unit (8) determines the resulting displacement of the object (4).

12. The method according to any one of Claims 1 to 11,
wherein a time filtering is performed on the sequence of the displacement data based on extended Kalman filtering for improving the accuracy of the estimation of the motion of the object (4).

13. The method according to any one of Claims 1 to 12,
wherein a object support system is utilized to position the object (4) in a predefined position with respect to the imaging system before starting the acquisition process.

14. The method according to any one of Claims 1 to 13,
wherein a reconstruction process resulting in tomographic images is used for the image processing.

15. The method according to Claim 14,
wherein the reconstruction process is based on an algebraic method for solving a system of equations connecting the absorption coefficients (a_j) associated with volume elements (16) of the object with projections values of the radiographic raw images taken at various positions along the trajectory (11) by calculating a weighted sum of the absorption elements (a_j) along radiation paths (14) associated with the projection values, the weighting coefficient (w_j) depending on the length of a segment of a given radiation path (14) lying within a given volume element (16), and wherein the weighting coefficients (w_j) are modified in dependency of the displacement data.

16. The method according to Claim 2 and 14,
wherein a frequency based reconstruction process is used, in which the absorption coefficient (a_j) associated with a given volume element (16) are obtained by applying a backprojection on the filtered radiographic raw images taken from various positions along the trajectory (11), wherein the backprojection is performed along the radiation paths associated with the imaging system on the virtual trajectory.

17. The method according to Claim 2 and 14,
wherein a frequency based reconstruction process is used, in which the absorption coefficient (a_j) associated with a given volume element (16) are obtained by applying a rebinning process on the radiographic raw images, in which the projection data of a two-dimensional slice of the object (4) are estimated on the basis of the displacement data, and by applying a two-dimensional reconstruction process on the projection data.

18. The method according to any one of Claims 1 to 17,
wherein the object (4) is a body of a patient and wherein an imaging system arranged for cone beam computer tomography, tomosynthesis, dental tomography is used.

19. The method according to any one of Claims 1 to 13,
wherein the object (4) is a body of a patient and wherein an imaging system arranged for dental panoramic radiography, cephalography or medical scanography is used.

20. A apparatus for radiography with X-rays, comprising:

- an imaging system comprising an X-ray source (2) and an X-ray detector (3) for generating radiographic raw images of an object (4);

- a sensor arrangement (6, 7) adapted for determining displacement data of the object (4);
- a control unit (8) adapted for controlling the motion of the imaging system and for generating views of the object (4) based on the radiographic raw images

characterized in that
the apparatus is arranged for performing a method according to any one of Claims 1 to 19.

FIG 1

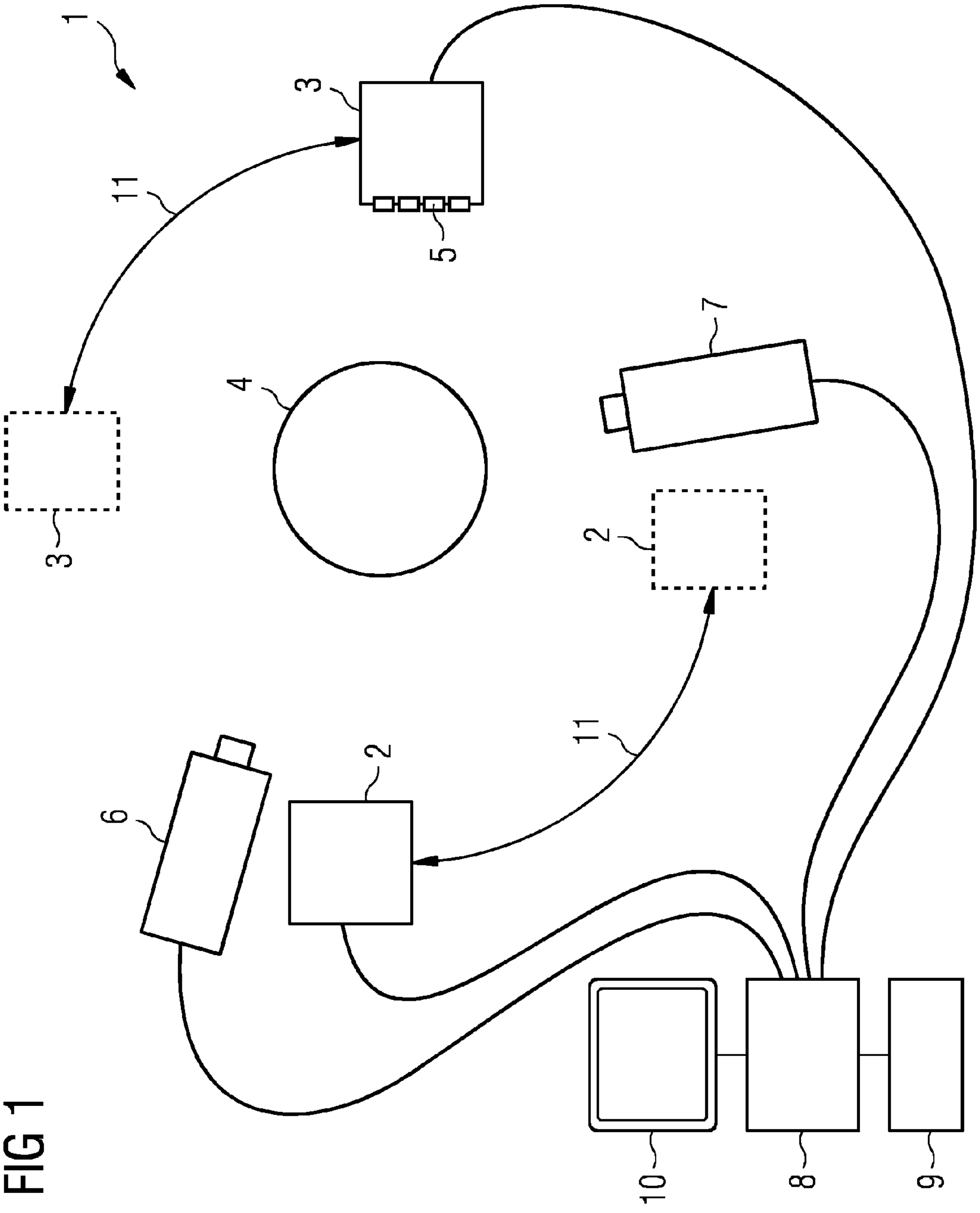


FIG 2

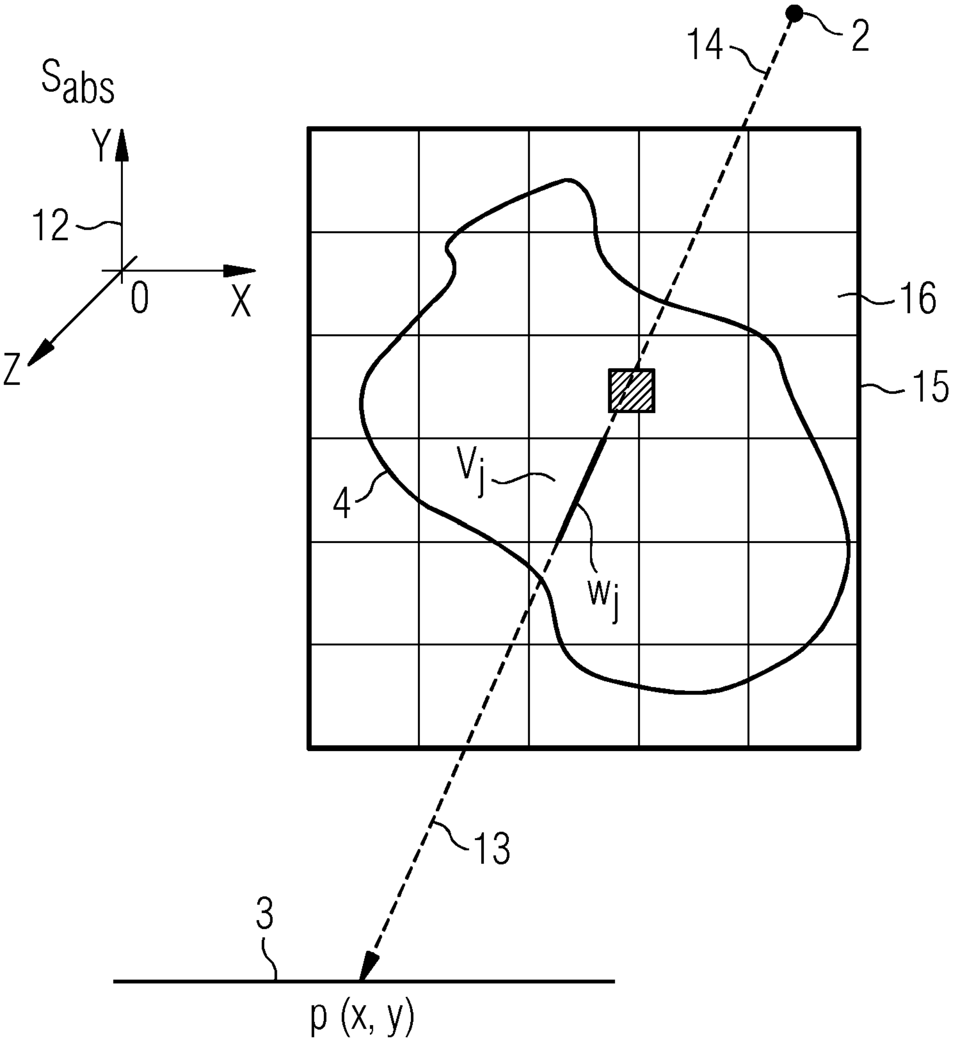


FIG 3

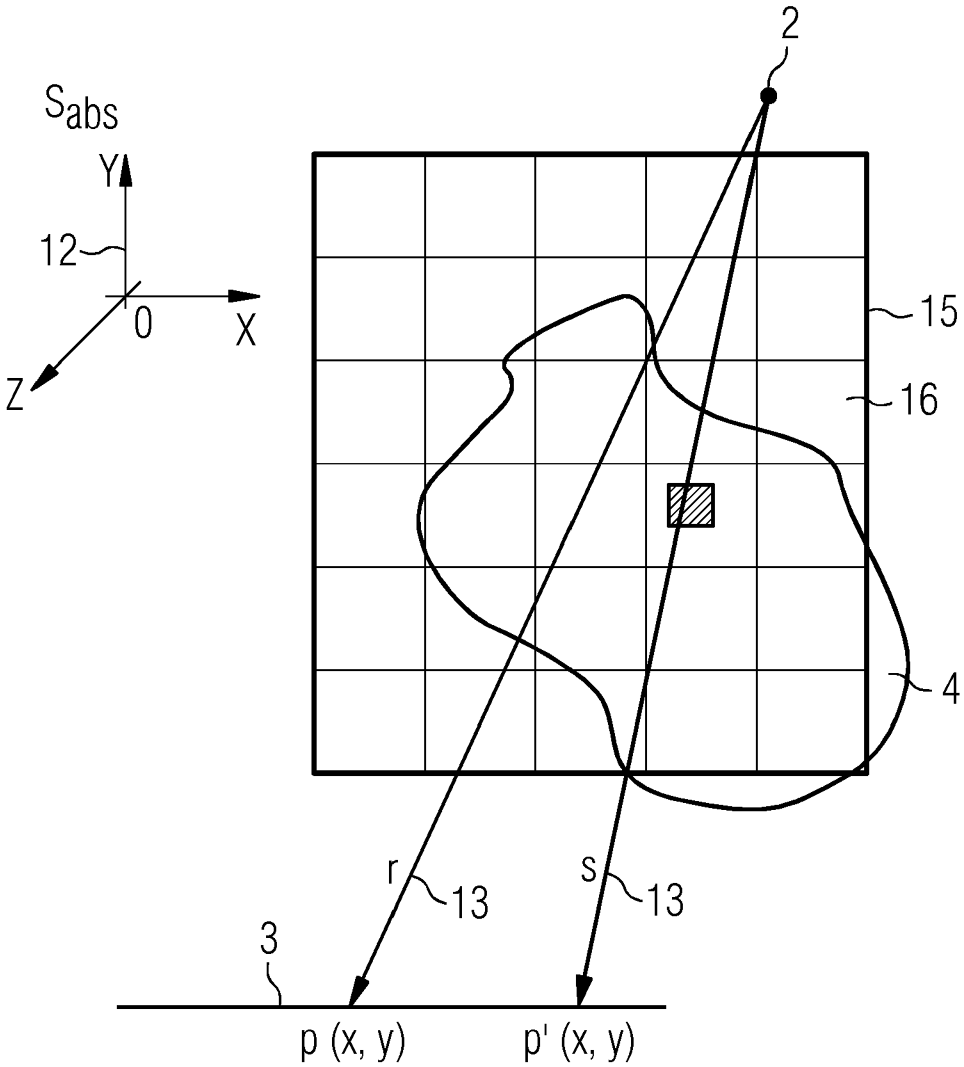
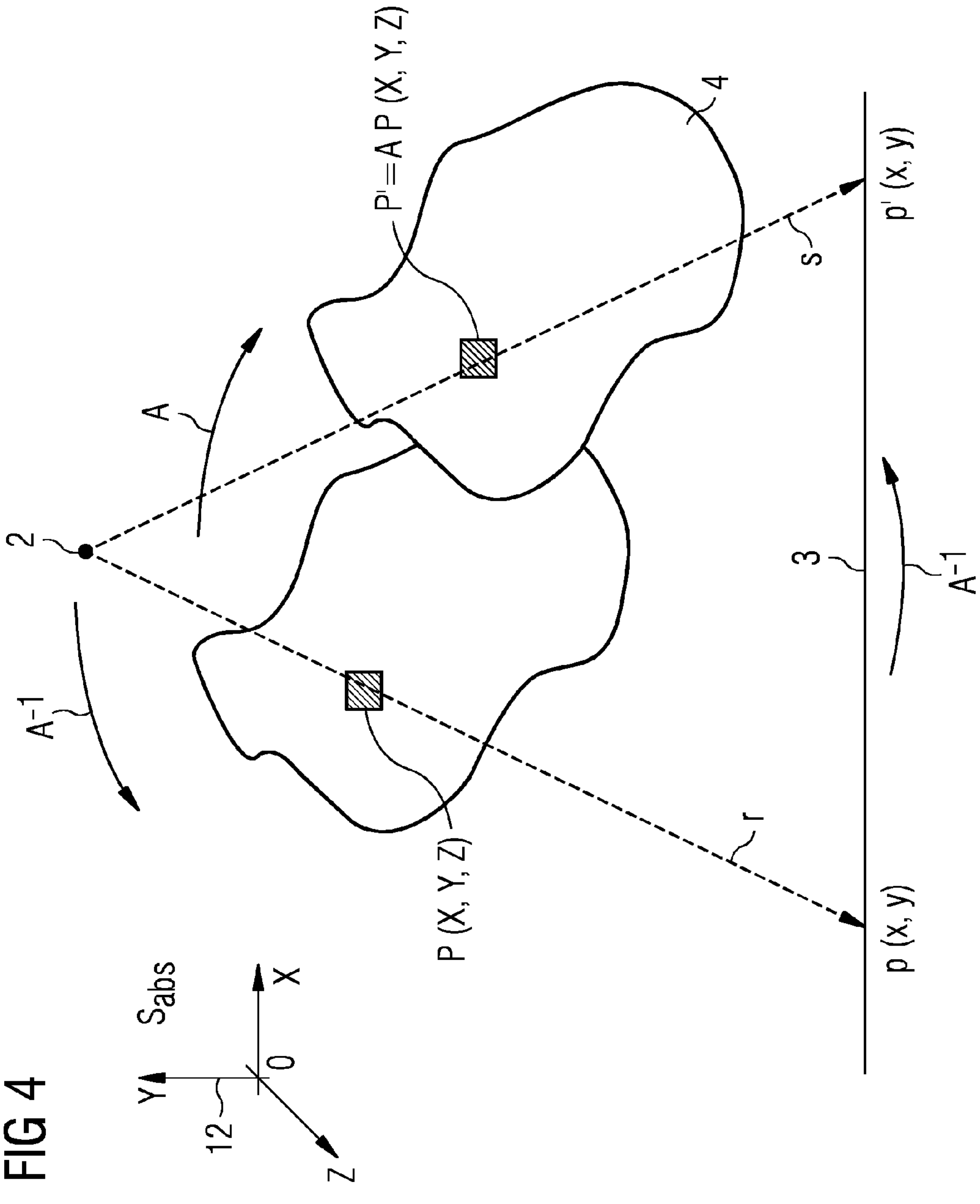


FIG 4



European Patent
Office

EUROPEAN SEARCH REPORT

Application Number
EP 08 16 0359

DOCUMENTS CONSIDERED TO BE RELEVANT

Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (IPC)
Y,D	US 7 187 749 B2 (SUZUKI MASAKAZU [JP]) 6 March 2007 (2007-03-06) * abstract; figure 1 * * column 3, line 39 - line 42 * * column 4, line 34 - line 41 * -----	1-20	INV. G06T11/00
Y	SÊBASTIEN ROUX ET AL: "Exact reconstruction in 2D dynamic CT: compensation of time-dependent affine deformations; Affine deformation compensation in 2D CT" PHYSICS IN MEDICINE AND BIOLOGY, TAYLOR AND FRANCIS LTD. LONDON, GB, vol. 49, no. 11, 7 June 2004 (2004-06-07), pages 2169-2182, XP020023724 ISSN: 0031-9155 * the whole document * -----	1-20	
A,D	EP 1 815 794 A (GENDEX DENTAL SYSTEMS S R L [IT]) 8 August 2007 (2007-08-08) * abstract; figure 1 * * column 1, paragraphs 6,7 * -----	1-20	TECHNICAL FIELDS SEARCHED (IPC)
A	LAURENT DESBAT ET AL: "Compensation of Some Time Dependent Deformations in Tomography" IEEE TRANSACTIONS ON MEDICAL IMAGING, IEEE SERVICE CENTER, PISCATAWAY, NJ, US, vol. 26, no. 2, 1 February 2007 (2007-02-01), pages 261-269, XP011161750 ISSN: 0278-0062 * abstract * -----	1-20	G06T

The present search report has been drawn up for all claims

Place of search	Date of completion of the search	Examiner
Munich	11 September 2008	Werling, Alexander

CATEGORY OF CITED DOCUMENTS

X : particularly relevant if taken alone
Y : particularly relevant if combined with another document of the same category
A : technological background
O : non-written disclosure
P : intermediate document

T : theory or principle underlying the invention
E : earlier patent document, but published on, or after the filing date
D : document cited in the application
L : document cited for other reasons
……………………………………………………………………
& : member of the same patent family, corresponding document

3

ANNEX TO THE EUROPEAN SEARCH REPORT ON EUROPEAN PATENT APPLICATION NO. EP 08 16 0359

This annex lists the patent family members relating to the patent documents cited in the above-mentioned European search report.
The members are as contained in the European Patent Office EDP file on
The European Patent Office is in no way liable for these particulars which are merely given for the purpose of information.

11-09-2008

Patent document cited in search report	Publication date	Patent family member(s)	Publication date
US 7187749 B2	06-03-2007	FI 20030621 A	25-10-2003
		JP 3785576 B2	14-06-2006
		JP 2004000568 A	08-01-2004
		US 2003215051 A1	20-11-2003
EP 1815794 A	08-08-2007	NONE	

EPO FORM P0459

For more details about this annex : see Official Journal of the European Patent Office, No. 12/82

REFERENCES CITED IN THE DESCRIPTION

Patent documents cited in the description

- US 7187749 B2 **[0003]**
- EP 1815794 A1 **[0006]**
- GB 2384155 A **[0012]**
- US 20060149134 A1 **[0013]**

Non-patent literature cited in the description

- **KAK ; SLANEY.** Principle of Computerized Tomographic Imaging. SIAM Society, 2001 **[0007]**
- **KALEDENDER, W. A.** X-ray computed tomography. *Phys. Med. Biol.,* 2006, vol. 51, R29-R43 **[0008]**
- **MICHEL, D. ; FRÉDÉRIC ; HIROYUKI, K.** Two-dimensional rebinning of helical cone-beam computerized tomography data using John's equation. *Inverse Problems,* 2003, vol. 19 (6), S41-S54 **[0009]**
- **DOBBINS III, J.T. ; GODFREY, D.J.** Digital X-ray tomosynthesis: current state of the art and clinical potential. *Phys. Med. Biol.,* 2003, vol. 48, R65-R106 **[0010]**
- **SHEPP, L. A. ; VARDI Y.** Maximum Likelihood Reconstruction for Emission Tomography. *IEEE Trans. Med. Im.,* 1982, vol. 1, 113-122 **[0011]**
- **VENUTO, F. ; LA CAMERA, A. et al.** Study of an iterative method for the reconstruction of images corrupted by Poisson and Gauss noise. *Inverse Problems,* 2008, 24 **[0014]**
- Multibiometric Systems, Comm. **JAIN, A.K. ; ROSS, A.** Special Issue on Multimodal Interfaces. ACM, 2004, vol. 47, 34-40 **[0015]**
- Face recognition by elastic bunch graph matching. **WISKOTT, J. et al.** Intelligent biometric techniques in fingerprints and face recognition. CRC Press, 1999, 355-396 **[0016]**
- **FERIS, R.S. et al.** Hierarchical wavelet networks for facial feature localization. *IEEE International Conference on Automatic Face and Gesture Recognition,* 2002, 118-123 **[0017]**
- **MIDDENDORFF, C. ; BOWEYER, K. ; YAN, P.** Multi-Modal Biometrics Involving the Human Ear. *Proc. IEEE Conf. CVPR 07,* 2007 **[0018]**
- **MA, Y. et al.** Robust precise eye location under probabilistic framework. *IEEE International Conference on Automatic Face and Gesture Recognition,* 2004, 339-344 **[0019]**
- **DARGHAM, J. ; CHEKIMA, A.** Lips Detection in the Normalised RGB Colour Scheme Proc. 2nd Conf. on Information and Communication technologies. *ICCTA,* 2006, 1546-15512006 **[0020]**
- **FROSIO, I. ; BORGHESE, N.A.** *Real-time accurate circle fitting with occlusions,* 2008, vol. 41, 101-114 **[0021]**
- **SCHWEIGER, M. ; ARRIDGE, S.R.** Image reconstruction in optical tomography, using local basis functions. *J. Electronic Imaging,* 2003, vol. 12 (4), 583-593 **[0053]**
- **HARTLEY, R. ; ZISSERMAN, A.** Multiple view geometry in Computer Vision. Cambridge University Press, 2004 **[0068]**