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(54) **Title**: MR IMAGING WITH B_1 MAPPING

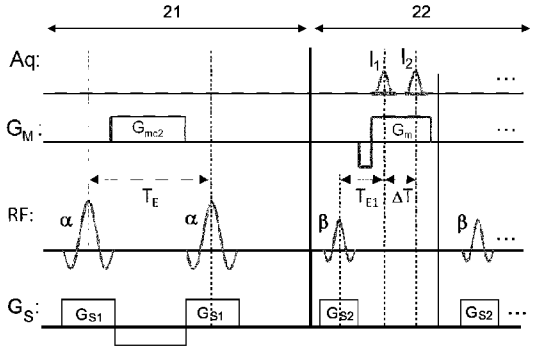


Fig. 2

(57) **Abstract**: The invention relates to a method of MR imaging of an object (10), wherein the object (10) to an imaging sequence (IMG) of RF pulses and switched magnetic field gradients, which imaging sequence (IMG) is a stimulated echo sequence including: i) at least two preparation RF pulses (a) radiated toward the object (10) in the presence of a slice-selection magnetic field gradient (GS) during a preparation period (21), and ii) one or more reading RF pulses (β) radiated toward the object (10) in the presence of a slice-selection magnetic field gradient (GS) during an acquisition period (22) temporally subsequent to the preparation period (21), wherein the slice-selection magnetic field gradients (GS) applied during the preparation period (21) and during the acquisition period (22) are of equal strength. One or more FID signals (Ii) as well as one or more stimulated echo signals (I2) are acquired during the acquisition period (22). At least one Bi map indicating the spatial distribution of the RF field of the RF pulses within object (10) is derived from the acquired FID (I_1) and stimulated echo (I_2) signals. Moreover, the invention relates to a MR device (1) and to a computer program for a MR device (1).

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MR imaging with B_1 mapping

FIELD OF THE INVENTION

The invention relates to the field of magnetic resonance (MR) imaging. It concerns methods of MR imaging of at least a portion of a body. The invention also relates to a MR device and to a computer program to be run on a MR device.

BACKGROUND OF THE INVENTION

Image-forming MR methods which utilize the interaction between magnetic fields and nuclear spins in order to form two-dimensional or three-dimensional images are widely used nowadays, notably in the field of medical diagnostics, because for the imaging of soft tissue they are superior to other imaging methods in many respects, do not require ionizing radiation and are usually not invasive.

According to the MR method in general, the body of the patient to be examined is arranged in a strong, uniform magnetic field (B_0 field) whose direction at the same time defines an axis (normally the z-axis) of the coordinate system on which the measurement is based. The magnetic field produces different energy levels for the individual nuclear spins in dependence on the magnetic field strength. These energy levels can be excited (spin resonance) by application of an electromagnetic alternating field (RF field, also referred to as B_1 field) of defined frequency (so-called Larmor frequency, or MR frequency). From a macroscopic point of view the distribution of the individual nuclear spins produces an overall magnetization which can be deflected out of the state of equilibrium by application of an electromagnetic pulse of appropriate frequency (RF pulse) while the magnetic field extends perpendicular to the z-axis, so that the magnetization performs a precessional motion about the z-axis. The precessional motion describes a surface of a cone whose angle of aperture is referred to as flip angle. The magnitude of the flip angle is dependent on the strength and the duration of the applied electromagnetic pulse. In the case of a so-called 90° pulse, the spins are deflected from the z axis to the transverse plane (flip angle 90°).

After termination of the RF pulse, the magnetization relaxes back to the original state of equilibrium, in which the magnetization in the z direction is built up again with a first time constant T_1 (spin lattice or longitudinal relaxation time), and the

magnetization in the direction perpendicular to the z direction relaxes with a second time constant T_2 (spin-spin or transverse relaxation time). The variation of the magnetization can be detected by means of one or more receiving RF coils which are arranged and oriented within an examination volume of the MR device in such a manner that the variation of the magnetization is measured in the direction perpendicular to the z-axis. The decay of the transverse magnetization is accompanied, after application of, for example, a 90° pulse, by a transition of the nuclear spins (induced by local magnetic field inhomogeneities) from an ordered state with the same phase to a state in which all phase angles are uniformly distributed (dephasing). The dephasing can be compensated by means of a refocusing pulse (for example a 180° pulse). This produces an echo signal (spin echo) in the receiving coils.

In order to realize spatial resolution in the body, linear magnetic field gradients extending along the three main axes are superposed on the uniform magnetic field, leading to a linear spatial dependency of the spin resonance frequency. The signal picked up in the receiving coils then contains components of different frequencies which can be associated with different locations in the body. The MR signal data obtained via the RF coils corresponds to the spatial frequency domain and is called k-space data. The k-space data usually includes multiple lines acquired with different phase encoding. Each line is digitized by collecting a number of samples. A set of k-space data is converted to a MR image by means of Fourier transformation.

It is generally desirable to have a relatively uniform homogeneity of the generated transmit RF field (B_1 field) for excitation of magnetic resonance throughout a cross section and/or a volume of the imaged patient's body. However, as the MR frequency increases, with increasing main magnetic field strength, this becomes more difficult due to conductive losses and wavelength effects within the body of the patient. Consequently, an accurate measurement of the spatial distribution of the transmitted RF field is important for many MR imaging applications to support appropriate prospective (if applicable) and retrospective correction/compensation. This requires a robust and fast B_1 mapping technique.

Fast and robust in vivo B_1 mapping is an essential pre-requisite for multi-coil transmit applications like RF shimming or accelerated multi-dimensional RF pulses. However, most B_1 mapping techniques are relatively slow, making integration into the clinical workflow difficult.

The recently introduced DREAM B_1 mapping approach (Magnetic Resonance in Medicine, 68, 1517-1526, 2012) allows the acquisition of a B_1 map in a short time interval. The DREAM approach includes a stimulated echo-type preparation phase in which locally

effective flip angles of the RF pulses of the preparation phase are encoded into longitudinal magnetization. This flip angle information is obtained in a subsequent acquisition phase in which a FID signal and a stimulated echo signal are acquired. A B_1 map indicating the spatial distribution of the local flip angle (i.e. the local amplitude of the RF pulses of the preparation phase) is then derived from the amplitudes of the acquired FID and stimulated echo signals.

In case of strong off-resonances, like, for example, B_0 variations caused by non-ideal shim settings, or by magnetic implants in the vicinity of the field of view, or strong chemical shifts, which approach or even exceed the bandwidth of the RF excitation pulses of the DREAM sequence, the MR excitation will degrade. This may cause a non-ideal flip angle-encoding, which will result accordingly in wrong estimates of the spatial distribution of the B_1 field.

SUMMARY OF THE INVENTION

From the foregoing it is readily appreciated that there is a need for an improved B_1 mapping method. It is thus an object of the invention to provide a refined DREAM sequence which allows proper B_1 mapping even in the presence of strong B_0 inhomogeneity.

In accordance with the invention, a method of MR imaging of an object placed in the examination volume of a MR device is disclosed. The method comprises the steps of:

- subjecting the object to an imaging sequence of RF pulses and switched magnetic field gradients, which imaging sequence is a stimulated echo sequence including:
  - i) at least two preparation RF pulses radiated toward the object in the presence of a slice-selection magnetic field gradient during a preparation period, and
  - ii) one or more reading RF pulses radiated toward the object in the presence of a slice-selection magnetic field gradient during an acquisition period temporally subsequent to the preparation period, wherein the slice-selection magnetic field gradients applied during the preparation period and during the acquisition period are of equal strength;
- acquiring one or more FID signals (I_1) and one or more stimulated echo signals (I_2) during the acquisition period (22); and
- deriving at least one B_1 map indicating the spatial distribution of the RF field of the RF pulses within object (10) from the acquired FID (I_1) and stimulated echo (I_2) signals.

In accordance with the invention, like in the known DREAM approach, one or more reading RF pulses are applied during the acquisition period of the stimulated echo

sequence, wherein the FID signals and the stimulated echo signals are acquired quasi-simultaneously. A MR image can be reconstructed from the FID signals and another MR image can be reconstructed from the stimulated echo signals. After the MR image reconstruction, the B_1 map can be derived from the voxel-wise intensity ratio of the two MR images reconstructed from the FID and stimulated echo signals, respectively.

A plurality of FID signals and stimulated echo signals with appropriate phase encoding need to be acquired for generating a complete B_1 map. Efficient sampling schemes like EPI, parallel imaging or compressed sensing can be advantageously applied for this purpose in combination with the invention.

With the slice-selective preparation RF pulses, an off-resonance (Larmor frequency shift ν_{off}) caused by strong B_0 inhomogeneity results in a corresponding shift Δx_{PREP} of the excited slice according to:

$$\Delta x_{PREP} = 2\pi\nu_{off} / \gamma G_{S,PREP},$$

wherein $G_{S,PREP}$ denotes the strength of the slice-selection magnetic field gradient applied during the preparation period. γ denotes the gyro-magnetic ratio. In the conventional DREAM technique, $G_{S,PREP}$ is typically several times smaller than the strength of the slice-selection magnetic field gradient applied during the acquisition period. This is due to the different bandwidths of the rather strong preparation RF pulses and the rather weak reading RF pulses. Therefore, the slice encoded by the preparation RF pulses and the slice encoded by the reading RF pulses will shift non-congruently in the presence of B_0 inhomogeneity. This results in the above mentioned problems, such as wrong estimates of the spatial distribution of the B_1 field.

According to the invention, in contrast, the slice-selection magnetic field gradients applied during the preparation period and during the acquisition period are of equal strength, resulting in a congruent shift of the slices excited during the preparation period and the acquisition period respectively. Hence, the B_1 encoding mechanism of the method remains valid, yielding proper B_1 maps. In practice, the off-resonance will result in a small spatial shift of the derived B_1 map. Considering the smoothness of typical B_1 maps, this can be expected to be a minor problem.

According to a preferred embodiment of the invention, the proposed approach can be used for parallel transmit applications, wherein the RF pulses are radiated toward the

portion of the body via two or more RF coils (or two or more sets of RF coils) in parallel. A B_1 map may be derived from the acquired FID and stimulated echo signals for each RF coil or set of RF coils, each B_1 map indicating the spatial distribution of the RF field of the RF pulses irradiated via the respective RF coil or set of RF coils. In this embodiment of the invention, B_1 mapping is performed for multiple RF transmit elements (RF coils) of the used MR device to map their corresponding transmit sensitivities. An individual mapping scan may be performed according to the method of the invention for each individual RF transmit element (or for a combination of them). A small spatial distortion of the B_1 map caused by a small spatial shift of the derived B_1 map (see above) is identical for each RF coil, and, hence, does not affect the RF shimming optimization problem.

The B_1 maps obtained according to the invention can advantageously be used in subsequent imaging scans for RF shimming. RF shimming refers to the spatial homogenization of the RF transmit field by adjustment of the complex amplitudes of the RF pulses radiated via the individual RF coils of a parallel transmit MR imaging system. Simple known RF shimming approaches use predefined, anatomy-specific RF shim sets, without taking the individual patient anatomy into account. According to the invention, the RF shim settings can be optimized in a patient specific way. For this purpose, one or more B_1 maps are derived from the FID and stimulated echo signals in the above-described fashion and optimal RF shim sets can be derived.

According to another preferred embodiment of the invention, a plurality of FID and stimulated echo MR signals are generated by means of a plurality of consecutive reading RF pulses, each having a flip angle of less than 90°, preferably less than 45°, most preferably less than 30°. As already mentioned above, a train of reading RF pulses having small flip angles can be used to achieve a fast readout of multiple FID and stimulated echo signals. As short as possible echo times can be used in order to minimize T_2^* relaxation.

The method of the invention described thus far can be carried out by means of a MR device including at least one main magnet coil for generating a uniform steady magnetic field within an examination volume, a number of gradient coils for generating switched magnetic field gradients in different spatial directions within the examination volume, at least one RF coil for generating RF pulses within the examination volume and/or for receiving MR signals from a body of a patient positioned in the examination volume, a control unit for controlling the temporal succession of RF pulses and switched magnetic field gradients, and a reconstruction unit for reconstructing MR images from the received MR

signals. The method of the invention is preferably implemented by a corresponding programming of the reconstruction unit and/or the control unit of the MR device.

The methods of the invention can be advantageously carried out in most MR devices in clinical use at present. To this end it is merely necessary to utilize a computer program by which the MR device is controlled such that it performs the above-explained method steps of the invention. The computer program may be present either on a data carrier or be present in a data network so as to be downloaded for installation in the control unit of the MR device.

BRIEF DESCRIPTION OF THE DRAWINGS

The enclosed drawings disclose preferred embodiments of the present invention. It should be understood, however, that the drawings are designed for the purpose of illustration only and not as a definition of the limits of the invention. In the drawings:

Figure 1 schematically shows a MR device for carrying out the methods of the invention;

Figure 2 shows a schematic diagram illustrating an imaging sequence according to the invention.

DETAILED DESCRIPTION OF THE EMBODIMENTS

With reference to Figure 1, a MR device 1 is shown. The device comprises superconducting or resistive main magnet coils 2 such that a substantially uniform, temporally constant main magnetic field B_0 is created along a z-axis through an examination volume. The device further comprises a set of (1^{st}, 2^{nd}, and - where applicable - 3^{rd} order) shimming coils 2', wherein the current flow through the individual shimming coils of the set 2' is controllable for the purpose of minimizing B_0 deviations within the examination volume.

A magnetic resonance generation and manipulation system applies a series of RF pulses and switched magnetic field gradients to invert or excite nuclear magnetic spins, induce magnetic resonance, refocus magnetic resonance, manipulate magnetic resonance, spatially and otherwise encode the magnetic resonance, saturate spins, and the like to perform MR imaging.

Most specifically, a gradient pulse amplifier 3 applies current pulses to selected ones of whole-body gradient coils 4, 5 and 6 along x, y and z-axes of the examination volume. A digital RF frequency transmitter 7 transmits RF pulses or pulse

packets, via a send-/receive switch 8, to a body RF coil 9 to transmit RF pulses into the examination volume. A typical MR imaging sequence is composed of a packet of RF pulse segments of short duration which taken together with each other and any applied magnetic field gradients achieve a selected manipulation of nuclear magnetic resonance. The RF pulses are used to saturate, excite resonance, invert magnetization, refocus resonance, or manipulate resonance and select a portion of a body 10 positioned in the examination volume. The MR signals are also picked up by the body RF coil 9.

For generation of MR images of limited regions of the body 10 by means of parallel imaging, a set of local array RF coils 11, 12, 13 are placed contiguous to the region selected for imaging. The array coils 11, 12, 13 can be used to receive MR signals induced by body-coil RF transmissions. In parallel transmit applications, the array RF coils 11, 12, 13 may also be used for RF transmission, for example for the purpose of RF shimming.

The resultant MR signals are picked up by the body RF coil 9 and/or by the array RF coils 11, 12, 13 and demodulated by a receiver 14 preferably including a preamplifier (not shown). The receiver 14 is connected to the RF coils 9, 11, 12 and 13 via send-/receive switch 8.

A host computer 15 controls the current flow through the shimming coils 2' as well as the gradient pulse amplifier 3 and the transmitter 7 to generate any of a plurality of MR imaging sequences, such as echo planar imaging (EPI), echo volume imaging, gradient and spin echo imaging, fast spin echo imaging, and the like. For the selected sequence, the receiver 14 receives a single or a plurality of MR data lines in rapid succession following each RF excitation pulse. A data acquisition system 16 performs analog-to-digital conversion of the received signals and converts each MR data line to a digital format suitable for further processing. In modern MR devices the data acquisition system 16 is a separate computer which is specialized in acquisition of raw image data.

Ultimately, the digital raw image data is reconstructed into an image representation by a reconstruction processor 17 which applies a Fourier transform or other appropriate reconstruction algorithms, such like SENSE or SMASH. The MR image may represent a planar slice through the patient, an array of parallel planar slices, a three-dimensional volume, or the like. The image is then stored in an image memory where it may be accessed for converting slices, projections, or other portions of the image representation into appropriate format for visualization, for example via a video monitor 18 which provides a man-readable display of the resultant MR image.

Figure 2 shows a diagram illustrating an imaging sequence according to the invention. The depicted imaging sequence is a stimulated echo sequence which is subdivided into a preparation period 21 and an acquisition period 22. Two preparation RF pulses having a flip angle of α are applied during the preparation period 21 in the presence of a slice-selection magnetic field gradient G_{S1}. The two preparation RF pulses are separated by a time interval T_E. A de-phaser magnetic field gradient G_{mc2} is applied between the two preparation RF pulses. A sequence of reading RF pulses having flip-angle β are generated during the acquisition period 22 in the presence of a slice-selection gradient G_{S2}. An FID signal I_1 and a stimulated echo signal I_2 are acquired after each reading pulse as gradient-recalled echoes. Applied phase-encoding magnetic field gradients are not shown in Figure 2 for simplification.

Directly after the preparation sequence 21, the longitudinal magnetization is given by:

$$M_{z1} = \cos^2(\alpha) \cdot M_0$$
$$M_{z2} = \frac{1}{2}\sin^2(\alpha) \cdot M_0 \quad ,$$

wherein M_{z1} and M_{z2} denote the un-prepared (i.e. in-phase) and the stimulated echo-prepared (i.e. de-phased) longitudinal magnetization, respectively. Both the FID signal I_1 generated from M_{z1} and the stimulated echo signal I_2 generated from M_{z2} are acquired at different points in time T_{E1} and $T_{E1} + \Delta T$, respectively. The delay ΔT between the two echoes I_1, I_2 is determined by the relation:

$$\Delta T = A_{mc2} / G_m \quad ,$$

wherein A_{mc2} denotes the gradient-time area of the de-phaser gradient G_{mc2} and G_m denotes the strength of the readout magnetic field gradient. Neglecting T_1- and T_2-effects, the two acquired echo signals I_1 and I_2 are given by:

$$I_1 = S \cdot C(T_{E1}) \sin(\beta) M_{z1}$$
$$I_2 = S \cdot C(T_{E1} + \Delta T - T_E) \sin(\beta) M_{z2} \quad ,$$

wherein S represents a complex system constant, which is equal for both echo signals I_1 and I_2 and which is determined e.g. by transmit and receive coil sensitivities for a given voxel. β is the nominal flip angle of the reading RF pulses. C describes the static signal de-phasing for a given voxel due to susceptibility and chemical shift effects:

$$C(t) = \int_V \rho(\mathbf{r}) e^{-i\omega(\mathbf{r})t} d\mathbf{r} \quad ,$$

wherein ρ and ω denote the proton density and the off-resonance frequency offset, respectively. The integral describes the summation over the given voxel. By applying the timing scheme

$$T_E = 2T_{E1} + \Delta T$$

the measured echo signals I_1 and I_2 are given by:

$$I_1 = S \cdot C(T_{E1}) \sin(\beta) M_{z1}$$
$$I_2 = S \cdot C^*(T_{E1}) \sin(\beta) M_{z2}$$

Thus, the de-phasing term *C* is identical for both echo signals, apart from the mirrored phase. For example by selecting T_{E1}=2.3 ms at a main magnetic field strength of 3 Tesla, signal contributions from water spins and signal contributions from fat spins are essentially in phase for both echoes I_1, I_2. Combining the above equations yields:

$$|I_2/I_1| = \tan^2(\alpha)/2$$

Thus, the unknown flip angle α of the stimulated echo preparation RF pulses can be derived from the ratio of the acquired echo signals according to:

$$\alpha = \arctan \sqrt{2|I_2/I_1|}$$

The imaging sequence shown in Fig. 2 is actually a basic version of the known DREAM B_1 mapping sequence (Magnetic Resonance in Medicine, 68, 1517-1526, 2012). However, G_{S1} is typically several times smaller than G_{S2} in the conventional DREAM

technique. This implies that the slice encoded by the preparation RF pulses (α) and the slice encoded by the reading RF pulses (β) will shift non-congruently in the presence of B_0 inhomogeneity, which results in wrong estimates of the B_1 map. According to the invention, as illustrated in Figure 2, the slice-selection magnetic field gradients G_{S1} applied during the preparation period and the slice-selection magnetic field gradients G_{S2} applied during the acquisition period are all of equal strength. In this way, a congruent shift of the slices excited during the preparation period and during the acquisition period is achieved. The B_1 encoding mechanism of the method remains valid, yielding proper B_1 maps. The off-resonance may result in a small spatial shift of the derived B_1 map. Considering the smoothness of typical B_1 maps, this can be neglected in most practical cases.

CLAIMS:

1. Method of MR imaging of an object (10) placed in the examination volume of a MR device (1), the method comprising the steps of:

- subjecting the object (10) to an imaging sequence (IMG) of RF pulses and switched magnetic field gradients, which imaging sequence (IMG) is a stimulated echo sequence including:

i) at least two preparation RF pulses (α), in particular having equal flip angels, radiated toward the object (10) in the presence of a slice-selection magnetic field gradient (GS) during a preparation period (21), and

ii) one or more reading RF pulses (β) radiated toward the object (10) in the presence of a slice-selection magnetic field gradient (GS) during an acquisition period (22) temporally subsequent to the preparation period (21), wherein the slice-selection magnetic field gradients (GS) applied during the preparation period (21) and during the acquisition period (22) are of equal strength;

- acquiring one or more FID signals (I_1) and one or more stimulated echo signals (I_2) during the acquisition period (22); and

- deriving at least one B_1 map indicating the spatial distribution of the RF field of the RF pulses within object (10) from the acquired FID (I_1) and stimulated echo (I_2) signals.

2. Method of claim 1, wherein the B_1 map is derived from the voxel-wise intensity ratio of the FID (I_1) and stimulated echo (I_2) signals.

3. Method of claim 1 or 2, wherein the RF pulses are radiated toward the portion of the body (10) via two or more RF coils (9, 11, 12, 13), wherein the B_1 map indicates the spatial distribution of the RF field of the RF pulses radiated via the two or more RF coils (9, 11, 12, 13).

4. Method of claim 3, wherein RF shim settings are derived from the B_1 map, wherein the amplitudes and phases of the RF pulses radiated toward the object (10) via the two or more RF coils (9, 11, 12, 13) are controlled according to the RF shim settings.

5. Method of any one of claims 1-4, wherein a plurality of FID (I_1) and stimulated echo (I_2) MR signals are generated by means of a plurality of consecutive reading RF pulses (β).

6. MR device comprising at least one main magnet coil (2) for generating a uniform, steady magnetic field within an examination volume, a number of gradient coils (4, 5, 6) for generating switched magnetic field gradients in different spatial directions within the examination volume, at least one RF coil (9) for generating RF pulses within the examination volume and/or for receiving MR signals from an object (10) positioned in the examination volume, a control unit (15) for controlling the temporal succession of RF pulses and switched magnetic field gradients, and a reconstruction unit (17) for reconstructing MR images from the received MR signals, wherein the MR device (1) is arranged to perform the following steps:

- subjecting the object (10) to an imaging sequence (IMG) of RF pulses and switched magnetic field gradients, which imaging sequence (IMG) is a stimulated echo sequence including:

  i) at least two preparation RF pulses (α) radiated toward the object (10) in the presence of a slice-selection magnetic field gradient (GS) during a preparation period (21), and

  ii) one or more reading RF pulses (β) radiated toward the object (10) in the presence of a slice-selection magnetic field gradient (GS) during an acquisition period (22) temporally subsequent to the preparation period (21), wherein the slice-selection magnetic field gradients (GS) applied during the preparation period (21) and during the acquisition period (22) are of equal strength;

- acquiring one or more FID signals (I_1) and one or more stimulated echo signals (I_2) during the acquisition period (22); and

- deriving at least one B_1 map indicating the spatial distribution of the RF field of the RF pulses within object (10) from the acquired FID (I_1) and stimulated echo (I_2) signals.

7. Computer program to be run on a MR device (1), which computer program comprises instructions for:

- generating an imaging sequence (IMG) of RF pulses and switched magnetic field gradients, which imaging sequence (IMG) is a stimulated echo sequence including:

i) at least two preparation RF pulses (α) radiated in the presence of a slice-selection magnetic field gradient (GS) during a preparation period (21), and

ii) one or more reading RF pulses (β) radiated in the presence of a slice-selection magnetic field gradient (GS) during an acquisition period (22) temporally subsequent to the preparation period (21), wherein the slice-selection magnetic field gradients (GS) applied during the preparation period (21) and during the acquisition period (22) are of equal strength;

- acquiring one or more FID signals (I_1) and one or more stimulated echo signals (I_2) during the acquisition period (22); and

- deriving at least one B_1 map indicating the spatial distribution of the RF field of the RF pulses from the acquired FID (I_1) and stimulated echo (I_2) signals.

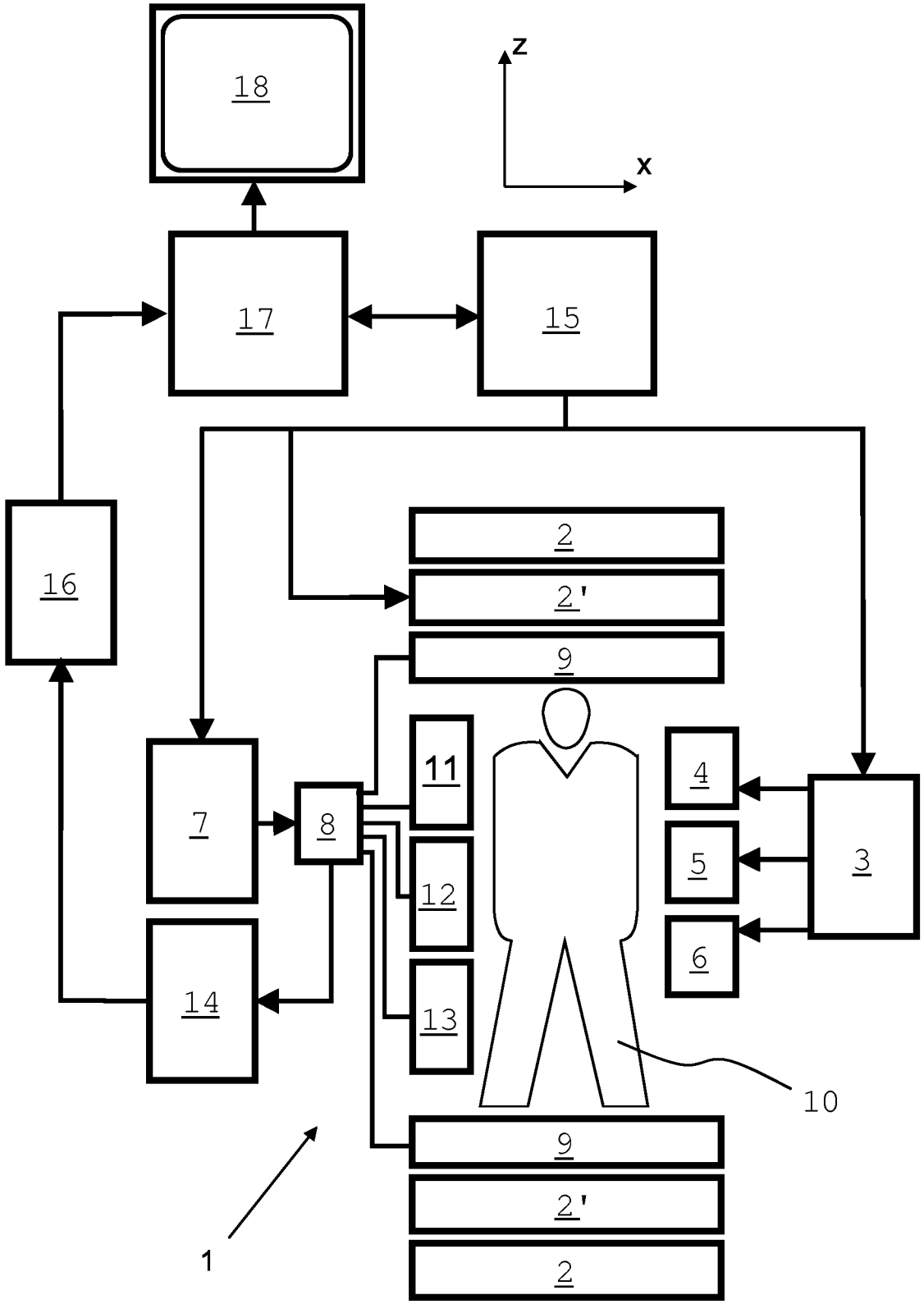


Fig. 1

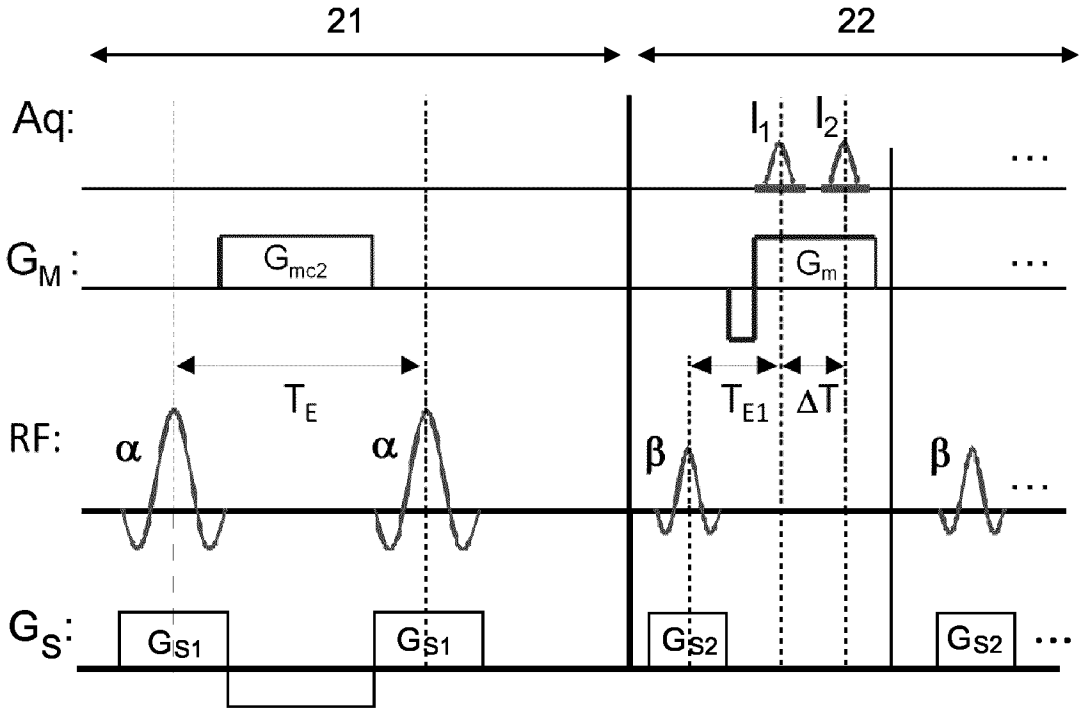


Fig. 2

INTERNATIONAL SEARCH REPORT

International application No
PCT/EP2016/051069

A. CLASSIFICATION OF SUBJECT MATTER
INV. G01R33/24
ADD.

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
G01R G01N G01V A61B

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

EPO-Internal, BIOSIS, COMPENDEX, EMBASE, INSPEC, IBM-TDB, WPI Data

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	AKOKA S ET AL: "Radiofrequency map of an NMR coil by imaging", MAGNETIC RESONANCE IMAGING, ELSEVIER SCIENCE, TARRYTOWN, NY, US, vol. 11, no. 3, 1 January 1993 (1993-01-01), pages 437-441, XP002466125, ISSN: 0730-725X, DOI: 10.1016/0730-725X(93)90078-R Sections "Principle", "Materials and Methods" figure 1	1-7
A	US 2012/139539 A1 (NITTKA MATHIAS [DE]) 7 June 2012 (2012-06-07) paragraph [0031] - paragraph [0036] figure 2	1-7

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[X] Further documents are listed in the continuation of Box C.

[X] See patent family annex.

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Date of the actual completion of the international search	Date of mailing of the international search report
3 May 2016	13/05/2016

Name and mailing address of the ISA/	Authorized officer
European Patent Office, P.B. 5818 Patentlaan 2 NL - 2280 HV Rijswijk Tel. (+31-70) 340-2040, Fax: (+31-70) 340-3016	Streif, Jörg Ulrich

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