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(54) **ARRANGEMENT AND METHOD FOR INFLUENCING AND/OR DETECTING MAGNETIC PARTICLES AND FOR MAGNETIC RESONANCE IMAGING**

ANORDNUNG UND VERFAHREN ZUM BEEINFLUSSEN UND/ODER DETEKTIEREN VON
MAGNETISCHEN TEILCHEN UND ZUR MAGNETRESONANZBILDGEBUNG

AGENCEMENT ET PROCÉDÉ POUR INFLUENCER ET/OU DÉTECTER DES PARTICULES
MAGNÉTIQUES ET POUR UNE IMAGERIE PAR RÉSONANCE MAGNÉTIQUE

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Description

FIELD OF THE INVENTION

[0001] The present invention relates to an arrangement and a method for influencing and/or detecting magnetic particles in a region of action and for magnetic resonance imaging of an examination object in said region of action. Further, the present invention relates to a computer program for implementing said method on a computer and to control such an arrangement.

BACKGROUND OF THE INVENTION

[0002] An arrangement for influencing and/or detecting magnetic particles in a region of action is known from German patent application DE 101 51 778 A1. In the arrangement described in that publication, first of all a magnetic selection field having a spatial distribution of the magnetic field strength is generated such that a first sub-zone having a relatively low magnetic field strength and a second sub-zone having a relatively high magnetic field strength are formed in the examination zone. The position in space of the sub-zones in the examination zone is then shifted, so that the magnetization of the particles in the examination zone changes locally. Signals are recorded which are dependent on the magnetization in the examination zone, which magnetization has been influenced by the shift in the position in space of the sub-zones, and information concerning the spatial distribution of the magnetic particles in the examination zone is extracted from these signals, so that an image of the examination zone can be formed. Such an arrangement has the advantage that it can be used to examine arbitrary examination objects - e. g. human bodies - in a non-destructive manner and without causing any damage and with a high spatial resolution, both close to the surface and remote from the surface of the examination object.

[0003] A similar arrangement and method is known from Gleich, B. and Weizenecker, J. (2005), "Tomographic imaging using the nonlinear response of magnetic particles" in nature, vol. 435, pp. 1214-1217. The arrangement and method for magnetic particle imaging (MPI) described in that publication takes advantage of the non-linear magnetization curve of small magnetic particles.

[0004] Arrangements and methods for magnetic resonance imaging (MRI) are widely known in the art. MR imaging has become one of the main imaging modalities, in particular in the medical field, and the details of the general layout of and MR system and the standard methods for MR imaging are described in many publications and standard books.

[0005] A combination of MPI and MRI would be very useful for a clinical acceptance of MPI. An MRI scan could be useful for planning the MPI examination or for identifying tissue parameters that are not accessible by MPI alone. For some patients, the scanner may be used as

a pure MRI scanner if no MPI examination is scheduled. International application WO2009/008956 A2 discloses a combined imaging modality where the particles are imaged using magnetic particle imaging and the anatomy is imaged using conventional magnetic resonance imaging. Such a hybrid imaging modality contains a first, a second and a third subsets of coils as defined in claim 1 of the appended claims, an excitation means for exciting magnetic resonance signals and a receiving means for acquiring magnetic resonance signals, a signal generation means for generating and providing field generation signals to said subsets of coils and processing means for processing the magnetic resonance signals and the magnetic particle imaging detection signals, respectively.

SUMMARY OF THE INVENTION

[0006] It is an object of the present invention to provide an arrangement and a method for influencing and/or detecting magnetic particles in a region of action and for magnetic resonance imaging of an examination object in said region of action, i.e. which allows to use the same hardware selectively for MPI data acquisition or MRI data acquisition or, in a combined mode, for data acquisition for both MPI and MRI.

[0007] In a first aspect of the present invention an arrangement for influencing and/or detecting magnetic particles in a region of action and for magnetic resonance imaging of an examination object in said region of action is presented in claim 1. For a combined arrangement (usable for MPI and MPI data acquisitions), as many MPI components as possible should be used for MRI. However, although MPI selection field coils can produce quite strong (0.4 ... 0.6 T) homogeneous fields, if a reverse current in one of the coil elements is applied compared to one or more of the other coil elements, the homogeneity is not good enough for an MRI operation. MPI usually generates RF fields useful for MRI only below 100 kHz, limiting the main field strength to a few mT. The receive coils typically operate below 2 MHz which limits the main field strength to 50 mT. One embodiment is thus based on the idea to use the MPI system as pre-polarized MRI system. This means that the protons are polarized by applying the strong homogeneous field for some time, e.g. several 100 ms. The RF pulses, spatial encoding and read-out of the MR signals is then done at a lower field strength.

[0008] The arrangement proposed according to the present invention particularly comprises a set of coils which is able to generate various (homogeneous and gradient) magnet fields, in particular a substantially homogeneous main magnetic field in a selected and changeable direction as well as a magnetic gradient field in at least two, preferably three, selected and changeable directions. The substantially homogeneous main magnetic field and the magnetic gradient field are generally almost stationary, but slowly varying (i.e. slowly compared to the

magnetic drive field as explained below). These magnetic fields can have frequency components up to approximately 1 kHz.

[0009] Preferably, the set of coils should be able to generate the field free point (FFP), i.e. first sub-zone having a low magnetic field strength, having a value of $40 \text{ T/sec} < \text{dB/dT} > 0.2 \text{ T/sec}$ and having a field strength of at least 100 mT. Further, it shall be able to generate a fast oscillating field homogenous (also called drive field) having a value of $\text{dB/dT} > 100 \text{ T/sec}$, preferably $> 1000 \text{ T/sec}$.

[0010] According to the present invention said set of coils comprises a number of subsets, in particular three subsets, wherein a first subset of coils is provided for generating said magnetic gradient field, a second subset of coils is provided for generating said temporally changing homogenous magnetic field for changing the position in space of the region of action, and a third subset of coils is provided for generating said temporally changing homogenous magnetic field for changing the position in space of the two sub-zones in the region of action so that the magnetization of the magnetic material changes locally. This provides sufficient flexibility to generate the desired magnetic fields depending on the mode of application. For this purpose, each subset is individually addressable and controllable, e.g. for each subset a separate generation means is separately provided for generating and providing field generation signals to said subsets of coils.

[0011] Using the terminology as conventionally used in a Magnetic Particle Imaging the first subset of coils comprises selection field coils for generating a magnetic selection field as said magnetic gradient field, said second subset of coils comprises focus field coils for generating a magnetic focus field as a temporally changing homogenous field for changing the position in space of the region of action, and said third subset of coils comprises drive field coils for generating a magnetic drive field as said temporally changing homogenous magnetic fields for changing the position in space of the two sub-zones in the region of action so that the magnetization of the magnetic material changes locally. In other words, the coils generally provided in an MPI system are advantageously used for generating the required magnetic fields in the MRI mode to acquire magnetic resonance signals. Thus, generally no additional hardware is required in the arrangement that can be used in both modes.

[0012] Preferably said substantially homogenous main magnetic field in a selected and changeable direction is generated by said first and/or second subset of coils and said magnetic gradient field in at least two selected and changeable directions is generated by said second subset of coils. This ensures an efficient use of the available subsets of coils without the need to add further coils for specific use in the MRI mode.

[0013] In an advantageous embodiment said first subset of coils (e.g. the selection field coils) comprises a

number of at least two, in particular a pair, of first coil elements located on opposite sides of the region of action, and said signal generation means are adapted for providing each first coil element with a separate first field generation signal. In other words, each of said first coil elements is separately controllable and providable with a first field generation signal, depending on which kind of magnetic field shall be generated therewith. For instance, in the MPI mode said first subset of coils is controlled such that a magnetic gradient field is generated having a pattern in space such that a first sub-zone having a low magnetic field strength and a second sub-zone having a higher magnetic field strength are formed in the region of action for which purpose, in case of oppositely arranged two coil elements, these coil elements are provided with oppositely oriented currents. In the MRI mode, however, both coil elements are provided with identically oriented currents to generate said substantially homogenous main magnetic field.

[0014] The second subset of coils preferably comprises a number of at least six, in particular three pairs, of second coil elements located on opposite sides of the region of action, wherein said signal generation means are adapted for providing each coil element with a separate second field generation signal. The coils of said second subset, which are preferably the focus field coils of the MPI system, can be used for generating the main magnetic field as well as the gradient fields in the MRI mode, depending on how they are provided with second field generation signals. In particular, if the coil elements of a pair of oppositely arranged coil elements are provided with oppositely oriented currents, a gradient field will be generated.

[0015] As mentioned above the arrangement according to the present invention can be used in various modes, in particular a magnetic particle imaging mode and a magnetic resonance imaging mode as well as a combined mode, in which particularly some waiting time required in the magnetic resonance imaging mode is used for obtaining magnetic particle imaging signals. The control means is accordingly adapted to switch between the various modes and to control the signal generation means accordingly so that the coils of the set of coils generate the required magnetic fields for the respective mode.

[0016] To provide a sufficient homogeneity for the main magnetic field in the MRI mode shimming coils are provided in an embodiment for generating a stationary and substantially homogenous main magnetic field in MRI mode. These shimming coils can be operated alone or in addition to the first and/or second subset of coils for generating the homogenous main field in the MRI mode, since the homogeneity provided by the first and/or second subset of coils might be too low for certain applications.

[0017] For exciting the magnetic spins in the region of action to precess in the MRI mode various options exist. According to an embodiment the control means is adapt-

ed for controlling said signal generation means to generate and provide field generation signals to said set of coils to subsequently

- generate a strong magnetic gradient field by said second subset of coils,
- reduce the gradient of said magnetic gradient field by said second subset of coils and add a substantially main homogeneous magnetic field by said first and/or second subset of coils, and
- reduce the field strength of said magnetic gradient field by said second subset of coils, and

wherein said control means is adapted to control said receiving means to then acquire magnetic resonance signals. According to another alternative embodiment said excitation means includes said third subset of coils which is adapted for generating RF transmit signals in a magnetic resonance imaging mode for exciting, in the magnetic resonance imaging mode, magnetic spins in the region of action to precess. The third subset of coils, can thus be used according to the present invention for generating the B_1 field in a magnetic resonance imaging mode, in particular if the main magnetic field (B_0) is made sufficiently small (e.g. 2.4 mT for 100 kHz). Further, they can be used when the main magnetic field is reduced to zero and then increased again with a different magnetization direction as explained above. Between these ways of use there is a smooth transition. If the field intensity of the drive field coil reaches the field intensity of the B_0 field, the spins can be excited even if the resonance condition (i.e. drive field frequency equals the Larmor frequency of the B_0 field) is not fulfilled since the B_0 field changes continuously.

[0018] In a combined MPI/MRI mode said control means is adapted for controlling said signal generation means to generate and provide field generation signals.

[0019] According to a further aspect of the present invention a method for influencing and/or detecting magnetic particles in a region of action and for magnetic resonance imaging of an examination object in said region of action is presented in claim 7. Finally, the present invention is directed to a computer program in accordance with claim 8. It shall be understood that all claimed methods and arrangements as well as the claimed computer program may have similar and/or identical preferred embodiments as defined in the dependent claims.

BRIEF DESCRIPTION OF THE DRAWINGS

[0020] These and other aspects of the invention will be apparent from and elucidated with reference to the embodiment(s) described hereinafter. In the following drawings

Fig. 1 shows a schematic view of the principle layout of a magnetic particle imaging (MPI) arrangement, Fig. 2 shows an example of the field line pattern pro-

duced by an arrangement as shown in Fig. 1, Fig. 3 shows an enlarged view of a magnetic particle present in the region of action, Figs. 4a and 4b show the magnetization characteristics of such particles, Fig. 5 shows a block diagram of a convention MR apparatus, Fig. 6 shows a block diagram of an embodiment of the arrangement according to the present invention, and Fig. 7 shows a block diagram of another embodiment of the arrangement according to the present invention.

15 DETAILED DESCRIPTION OF THE INVENTION

[0021] Before the details of the present invention shall be explained, basics of magnetic particle imaging shall be explained in details with reference to Figs. 1 to 4.

[0022] Fig. 1 shows an arbitrary object to be examined by means of a MPI arrangement 10. The reference numeral 1 in Fig. 1 denotes an object, in this case a human or animal patient, who is arranged on a patient table 2, only part of the top of which is shown. Prior to the application of the imaging method, magnetic particles 100 (not shown in Fig. 1) are arranged in a region of action 300 of the inventive arrangement 10. Especially prior to a therapeutic and/or diagnostic treatment of, for example, a tumor, the magnetic particles 100 are positioned in the region of action 300, e.g. by means of a liquid (not shown) comprising the magnetic particles 100 which is injected into the body of the patient 1.

[0023] As an example of an arrangement of coils is shown in Fig. 2 comprising a plurality of coils forming a selection means 210 whose range defines the region of action 300 which is also called the region of treatment 300. For example, the selection means 210 is arranged above and below the patient 1 or above and below the table top. For example, the selection means 210 comprise a first pair of coils 210', 210", each comprising two identically constructed windings 210' and 210" which are arranged coaxially above and below the patient 1 and which are traversed by equal currents, especially in opposed directions. The first coil pair 210', 210" together are called selection means 210 in the following. Preferably, direct currents are used in this case. The selection means 210 generate a magnetic selection field 211 which is in general a gradient magnetic field which is represented in Fig. 2 by the field lines. It has a substantially constant gradient in the direction of the (e.g. vertical) axis of the coil pair of the selection means 210 and reaches the value zero in a point on this axis. Starting from this field-free point (not individually shown in Fig. 2), the field strength of the magnetic selection field 211 increases in all three spatial directions as the distance increases from the field-free point. In a first sub-zone 301 or region 301 which is denoted by a dashed line around the field-free point the field strength is so small that the magnetization of parti-

cles 100 present in that first sub-zone 301 is not saturated, whereas the magnetization of particles 100 present in a second sub-zone 302 (outside the region 301) is in a state of saturation. The field-free point or first sub-zone 301 of the region of action 300 is preferably a spatially coherent area; it may also be a punctiform area or else a line or a flat area. In the second sub-zone 302 (i.e. in the residual part of the region of action 300 outside of the first sub-zone 301) the magnetic field strength is sufficiently strong to keep the particles 100 in a state of saturation. By changing the position of the two sub-zones 301, 302 within the region of action 300, the (overall) magnetization in the region of action 300 changes. By measuring the magnetization in the region of action 300 or a physical parameter influenced by the magnetization, information about the spatial distribution of the magnetic particles in the region of action can be obtained. In order to change the relative spatial position of the two sub-zones 301, 302 in the region of action 300, a further magnetic field, the so-called magnetic drive field 221, is superposed to the selection field 211 in the region of action 300 or at least in a part of the region of action 300.

[0024] Fig. 3 shows an example of a magnetic particle 100 of the kind used together with an arrangement 10 as shown above, but also according to the present invention. It comprises for example a spherical substrate 101, for example, of glass which is provided with a soft-magnetic layer 102 which has a thickness of, for example, 5 nm and consists, for example, of an iron-nickel alloy (for example, Permalloy). This layer may be covered, for example, by means of a coating layer 103 which protects the particle 100 against chemically and/or physically aggressive environments, e.g. acids. The magnetic field strength of the magnetic selection field 211 required for the saturation of the magnetization of such particles 100 is dependent on various parameters, e.g. the diameter of the particles 100, the used magnetic material for the magnetic layer 102 and other parameters.

[0025] In the case of e.g. a diameter of 10 μm , a magnetic field of approximately 800 A/m (corresponding approximately to a flux density of 1 mT) is then required, whereas in the case of a diameter of 100 μm a magnetic field of 80 A/m suffices. Even smaller values are obtained when a coating 102 of a material having a lower saturation magnetization is chosen or when the thickness of the layer 102 is reduced.

[0026] For further details of the preferred magnetic particles 100, reference can be made to the corresponding parts of DE 10151778, especially paragraphs 16 to 20 and paragraphs 57 to 61 of EP 1304542 A2 claiming the priority of DE 10151778.

[0027] The size of the first sub-zone 301 is dependent on the one hand on the strength of the gradient of the magnetic selection field 211 and on the other hand on the field strength of the magnetic field required for saturation. For a sufficient saturation of the magnetic particles 100 at a magnetic field strength of 80 A/m and a gradient (in a given space direction) of the field strength of the

magnetic selection field 211 amounting to $160 \cdot 10^3 \text{ A/m}^2$, the first sub-zone 301 in which the magnetization of the particles 100 is not saturated has dimensions of about 1 mm (in the given space direction).

[0028] When a further magnetic field - in the following called a magnetic drive field 221 is superposed on the magnetic selection field 210 (or gradient magnetic field 210) in the region of action 300, the first sub-zone 301 is shifted relative to the second sub-zone 302 in the direction of this magnetic drive field 221; the extent of this shift increases as the strength of the magnetic drive field 221 increases. When the superposed magnetic drive field 221 is variable in time, the position of the first sub-zone 301 varies accordingly in time and in space. It is advantageous to receive or to detect signals from the magnetic particles 100 located in the first sub-zone 301 in another frequency band (shifted to higher frequencies) than the frequency band of the magnetic drive field 221 variations. This is possible because frequency components of higher harmonics of the magnetic drive field 221 frequency occur due to a change in magnetization of the magnetic particles 100 in the region of action 300 as a result of the non-linearity of the magnetization characteristics.

[0029] In order to generate these magnetic drive fields 221 for any given direction in space, there are provided three further coil pairs, namely a second coil pair 220', a third coil pair 220" and a fourth coil pair 220''' which together are called drive means 220 in the following. For example, the second coil pair 220' generates a component of the magnetic drive field 221 which extends in the direction of the coil axis of the first coil pair 210', 210" or the selection means 210, i.e. for example vertically. To this end the windings of the second coil pair 220' are traversed by equal currents in the same direction. The effect that can be achieved by means of the second coil pair 220' can in principle also be achieved by the superposition of currents in the same direction on the opposed, equal currents in the first coil pair 210', 210", so that the current decreases in one coil and increases in the other coil. However, and especially for the purpose of a signal interpretation with a higher signal to noise ratio, it may be advantageous when the temporally constant (or quasi constant) selection field 211 (also called gradient magnetic field) and the temporally variable vertical magnetic drive field are generated by separate coil pairs of the selection means 210 and of the drive means 220.

[0030] The two further coil pairs 220", 220''' are provided in order to generate components of the magnetic drive field 221 which extend in a different direction in space, e.g. horizontally in the longitudinal direction of the region of action 300 (or the patient 1) and in a direction perpendicular thereto. If third and fourth coil pairs 220", 220''' of the Helmholtz type (like the coil pairs for the selection means 210 and the drive means 220) were used for this purpose, these coil pairs would have to be arranged to the left and the right of the region of treatment or in front of and behind this region, respectively. This

would affect the accessibility of the region of action 300 or the region of treatment 300. Therefore, the third and/or fourth magnetic coil pairs or coils 220', 220'' are also arranged above and below the region of action 300 and, therefore, their winding configuration must be different from that of the second coil pair 220'. Coils of this kind, however, are known from the field of magnetic resonance apparatus with open magnets (open MRI) in which an radio frequency (RF) coil pair is situated above and below the region of treatment, said RF coil pair being capable of generating a horizontal, temporally variable magnetic field. Therefore, the construction of such coils need not be further elaborated herein.

[0031] The arrangement 10 further comprises receiving means 230 that are only schematically shown in Fig. 1. The receiving means 230 usually comprise coils that are able to detect the signals induced by magnetization pattern of the magnetic particles 100 in the region of action 300. Coils of this kind, however, are known from the field of magnetic resonance apparatus in which e.g. a radio frequency (RF) coil pair is situated around the region of action 300 in order to have a signal to noise ratio as high as possible. Therefore, the construction of such coils need not be further elaborated herein.

[0032] In an alternative embodiment for the selection means 210 shown in Fig. 1, permanent magnets (not shown) can be used to generate the gradient magnetic selection field 211. In the space between two poles of such (opposing) permanent magnets (not shown) there is formed a magnetic field which is similar to that of Fig. 2, that is, when the opposing poles have the same polarity. In another alternative embodiment of the arrangement, the selection means 210 comprise both at least one permanent magnet and at least one coil 210', 210'' as depicted in Fig. 2.

[0033] The frequency ranges usually used for or in the different components of the selection means 210, drive means 220 and receiving means 230 are roughly as follows: The magnetic field generated by the selection means 210 does either not vary at all over the time or the variation is comparably slow, preferably between approximately 1 Hz and approximately 100 Hz. The magnetic field generated by the drive means 220 varies preferably between approximately 25 kHz and approximately 100 kHz. The magnetic field variations that the receiving means are supposed to be sensitive are preferably in a frequency range of approximately 50 kHz to approximately 10 MHz.

[0034] Figs. 4a and 4b show the magnetization characteristic, that is, the variation of the magnetization M of a particle 100 (not shown in Figs. 4a and 4b) as a function of the field strength H at the location of that particle 100, in a dispersion with such particles. It appears that the magnetization M no longer changes beyond a field strength $+H_c$ and below a field strength $-H_c$, which means that a saturated magnetization is reached. The magnetization M is not saturated between the values $+H_c$ and $-H_c$.

[0035] Fig. 4a illustrates the effect of a sinusoidal magnetic field $H(t)$ at the location of the particle 100 where the absolute values of the resulting sinusoidal magnetic field $H(t)$ (i.e. "seen by the particle 100") are lower than the magnetic field strength required to magnetically saturate the particle 100, i.e. in the case where no further magnetic field is active. The magnetization of the particle 100 or particles 100 for this condition reciprocates between its saturation values at the rhythm of the frequency of the magnetic field $H(t)$. The resultant variation in time of the magnetization is denoted by the reference $M(t)$ on the right hand side of Fig. 4a. It appears that the magnetization also changes periodically and that the magnetization of such a particle is periodically reversed.

[0036] The dashed part of the line at the centre of the curve denotes the approximate mean variation of the magnetization $M(t)$ as a function of the field strength of the sinusoidal magnetic field $H(t)$. As a deviation from this centre line, the magnetization extends slightly to the right when the magnetic field H increases from $-H_c$ to $+H_c$ and slightly to the left when the magnetic field H decreases from $+H_c$ to $-H_c$. This known effect is called a hysteresis effect which underlies a mechanism for the generation of heat. The hysteresis surface area which is formed between the paths of the curve and whose shape and size are dependent on the material, is a measure for the generation of heat upon variation of the magnetization.

[0037] Fig. 4b shows the effect of a sinusoidal magnetic field $H(t)$ on which a static magnetic field H_1 is superposed. Because the magnetization is in the saturated state, it is practically not influenced by the sinusoidal magnetic field $H(t)$. The magnetization $M(t)$ remains constant in time at this area. Consequently, the magnetic field $H(t)$ does not cause a change of the state of the magnetization.

[0038] In Fig. 5 the general layout of an MR imaging device, as for instance described in WO 2006/033047 A, is illustrated as a block diagram. The apparatus 100 comprises a set of main magnetic coils 102 for generating a stationary and homogeneous main magnetic field and three sets of gradient coils 103, 104 and 105 for superimposing additional magnetic fields with controllable strength and having a gradient in a selected direction. Conventionally, the direction of the main magnetic field is labelled the z-direction, the two directions perpendicular thereto the x- and y-directions. The gradient coils 103, 104, and 105 are energized via a power supply 111. The apparatus 100 further comprises an RF transmit antenna 106, for example a conventional quadrature body coil, for emitting RF pulses to a body 107. Being a quadrature coil, the antenna 106 has two resonance modes that are conventionally excited by RF signals having the same amplitudes and a phase difference of 90° . In accordance with the invention, the antenna 106 is connected to a receive/transmit unit 108 for controlling the phases and amplitudes of the two different resonance modes of the antenna 6. The receive/transmit unit 108 is further coupled to a modulator 109 for generating and modulat-

ing the RF pulses.

[0039] As shown in Fig. 5, the RF transmit antenna and the receiving antenna are physically the same antenna 106. Therefore, the transmit/receive unit 109 is arranged to separate the received signals from the RF pulses to be emitted. The received MR signals are input to a demodulator 110. The transmit/receive unit 108, the modulator 109, and the power supply 111 for the gradient coils 103, 104 and 105 are controlled by a control system 112. Control system 112 controls the phases and amplitudes of the RF signals fed to the antenna 106 on the basis of the aspect ratio of the body 300. The control system 112 is usually a microcomputer with a memory and a program control. The demodulator 110 is coupled to a data processing unit 114, for example a computer, for transformation of the received echo signals (also called magnetic resonance signals) into an image that can be made visible, for example, on a visual display unit 115.

[0040] Fig. 6 shows a block diagram of an embodiment of the apparatus 10 according to the present invention. The general principles of magnetic particle imaging and of magnetic resonance imaging explained above are valid and applicable here as well, unless otherwise specified.

[0041] The embodiment of the apparatus 10 shown in Fig. 5 comprises a set of various coils for generating the desired magnetic fields, desired on the mode of operation of the apparatus 10. First, the coils and their functions in a MPI mode shall be explained.

[0042] For generating the magnetic selection field explained above selection means are provided comprising a subset of selection field (SF) coils 23, preferably comprising a pair of coil elements, for generating said magnetic gradient field. The selection means further comprises a selection field signal generator unit 20. Preferably, a separate generator subunit is provided for each coil element of the subset 23 of selection field coils. Said selection field signal generator unit 20 comprises a controllable selection field current source 21 (generally including an amplifier) and a filter unit 22 which provide the respective section field coil element with the selection field current to individually set the gradient strength of the selection field in the desired direction. Preferably, a DC current is provided. If the selection field coil elements are arranged as opposed coils, e.g. on opposite sides of the region of action, the selection field currents of opposed coils are preferably oppositely oriented.

[0043] The selection field signal generator unit 20 is controlled by a control unit 70, which preferably controls the selection field current generation such that the sum of the field strength and the sum of the gradient strength of all spatial fractions of the selection field is maintained at a predefined level.

[0044] For generation of a magnetic focus field the apparatus 10 further comprises focus means comprising a subset of focus field (FF) coils 33a, 33b, 33c, preferably comprising three pairs 33a, 33b, 33c of oppositely ar-

anged focus field coil elements. Said magnetic focus field is generally used for changing the position in space of the region of action as is generally known in the art of MPI. The apparatus 10 further comprises a focus field signal generator unit 30 comprising a focus field current source 31 (preferably comprising a current amplifier) and a filter unit 32 for providing a focus field current to the respective coil of said subset of coils 33a, 33b, 33c which shall be used for generating the magnetic focus field. The focus field current source 31 is also controlled by the control unit 70.

[0045] For generation of the magnetic drive field the apparatus 10 further comprises drive means comprising a subset of drive field (DF) coils 43a, 43b, 43c, preferably comprising three pairs 43a, 43b, 43c of oppositely arranged drive field coil elements. Said magnetic drive field is generally used as said temporally changing homogeneous magnetic field for changing the position in space of the two sub-zones in the region of action so that the magnetization of the magnetic material changes locally. The apparatus 10 further comprises a drive field signal generator unit 40, preferably a separate drive field signal generation unit for each coil element of said set subset of drive field coils 43a, 43b, 43c. Said drive field signal generator unit 40 comprises a drive field current source 41 (preferably including a current amplifier) and a filter unit 42 for providing a drive field current to the respective drive field coil. The drive field current source 41 is adapted for generating an AC current and is also controlled by the control unit 70.

[0046] For signal detection receiving means 54, in particular a receiving coil, and a signal receiving unit 50, which receives signals detected by said receiving means 54, are provided. Said signal receiving unit 50 comprises a filter unit 51 for filtering the received detection signals. The aim of this filtering is to separate measured values, which are caused by the magnetization in the examination area which is influenced by the change in position of the two part-regions (301, 302), from other, interfering signals. To this end, the filter unit 51 may be designed for example such that signals which have temporal frequencies that are smaller than the temporal frequencies with which the receiving coil 54 is operated, or smaller than twice these temporal frequencies, do not pass the filter unit 51. The signals are then transmitted via an amplifier unit 52 to an analog/digital converter 53 (ADC). The digitalized signals produced by the analog/digital converter 53 are fed to an image processing unit (also called reconstruction means) 71, which reconstructs the spatial distribution of the magnetic particles from these signals and the respective position which the first part-region 301 of the first magnetic field in the examination area assumed during receipt of the respective signal and which the image processing unit 71 obtains from the control unit 70. The reconstructed spatial distribution of the magnetic particles is finally transmitted via the control means 70 to a computer 12, which displays it on a monitor 13. Thus, an image can be displayed showing the distri-

bution of magnetic particles in the examination area.

[0047] Further, an input unit 14 is provided, for example a keyboard. A user is therefore able to set the desired direction of the highest resolution and in turn receives the respective image of the region of action on the monitor 13. If the critical direction, in which the highest resolution is needed, deviates from the direction set first by the user, the user can still vary the direction manually in order to produce a further image with an improved imaging resolution. This resolution improvement process can also be operated automatically by the control unit 70 and the computer 12. The control unit 70 in this embodiment sets the gradient field in a first direction which is automatically estimated or set as start value by the user. The direction of the gradient field is then varied stepwise until the resolution of the thereby received images, which are compared by the computer 12, is maximal, respectively not improved anymore. The most critical direction can therefore be found respectively adapted automatically in order to receive the highest possible resolution.

[0048] According to the present invention the arrangement as illustrated in Fig. 5, which is generally designed for magnetic particle imaging is adapted such that it is also usable for magnetic resonance imaging. In particular, the set of coils provided in the MPI arrangement are controlled and provided with field generating currents such that the magnetic fields required for MRI can be generated as will be explained below.

[0049] The set of coils is particularly able to generate a substantially homogeneous main magnetic field in a selected and changeable direction, for which purpose preferably the selection field coils 23 and/or one or more of the focus field coils 33a, 33b, 33c are used. The set of coils is further able to generate a magnetic gradient field in at least two, preferably three orthogonal, selected and changeable directions, for which purpose preferably the focus field coils 33a, 33b, 33c are used. Further, an excitation means is provided for exciting magnetic spins in the region of action to precess, for which various embodiments exist as will be explained in detail below.

[0050] It shall be noted that the subsets of coils preferably comprise pairs of oppositely arranged coil elements. But it is also possible that instead of pairs more than two coil elements for generating the desired magnetic fields, e.g. three or four coil elements, are used. Further, according to an embodiment of the present invention all coils elements of the selection field coils 23 and of the focus field coils 33a, 33b, 33c are provided with a separate generator sub-unit (which may be part of a common generator unit) to individually control each coil element for generation of the desired magnetic field. The drive field coils 43a, 43b, 43c can have integrated receive coils and the focus field coils 33a, 33b, 33c can also serve as the selection field coils 23.

[0051] The control unit 70 is adapted for controlling the various generator unit to work in a MPI mode, a MRI mode or in a combined MPI/MRI mode, i.e. to switch between said mode (in particular in response to a user com-

mand) and to generate the respective control signals for the respective coils.

[0052] According to an aspect of the present invention it is proposed to use the MPI system as pre-polarized MRI system. This means that the protons are polarized by applying the strong homogeneous field for several 100 ms. The RF pulses, spatial encoding and read-out is done at a lower field strength. Such a pre-polarized MRI system is described by Macovski A, Conolly S, "Novel approaches to low-cost MRI", Magn. Reson. Med. 1993 Aug. 30, (2):221-30. In Prepolarized MRI (PMRI) with pulsed electromagnets, the main magnetic field cycles between a high field (B_p) to polarize the sample and a homogeneous, low field (B_0) for data acquisition. This architecture combines the higher SNR of the polarizing field with the imaging benefits of the lower field. However, PMRI can only achieve high SNR efficiency for volumetric imaging with 3D rapid imaging techniques, such as rapid acquisition with relaxation enhancement (RARE) (FSE, TSE), because slice- interleaved acquisition and longitudinal magnetization storage are both inefficient in PMRI.

[0053] There are several ways to use the MPI system as MRI system. In a first embodiment as illustrated in Fig. 7 shimming coils 83 (and a corresponding shimming coil signal generator unit 80 including a current source 81 and a filter 82) are added to generate a low field strength magnetic field with sufficient homogeneity for MRI operation. Further, an RF sender 63, in particular an RF transmit coil 63, is added coupled to a transmit signal generator unit 60 (including a send amplifier 61 and a filter 62) to generate the necessary RF field for excitation. The system may operate at about 50 mT.

[0054] In a second embodiment, as shown in Fig. 6, the transmit signal generator unit 60 is coupled to the receiver means 54, i.e. the receive coil, for generating the necessary RF field.

[0055] In a third embodiment only the shimming coils 83 and the shimming coil signal generator unit 80 are added (but no transmit signal generator unit 60 and RF sender 63). Hence, there is no RF B_1 field at the receive frequency available. The RF pulse is generated by cycling the main field down to a level, where the drive field frequency is the Larmor frequency. Receiving is thereafter done after ramping to e.g. 50 mT. Alternatively, the RF pulse is generated by decreasing the main magnetic field down to zero field and increasing the main field thereafter using a different magnetization direction. In particular, the main magnetic field is generated again in a direction, which does not correspond to the present magnetization direction, which generally corresponds to the direction of the main magnetic field that had existed shortly before reaching zero value when decreasing the main magnetic field.

[0056] In a fourth embodiment neither any shimming coils 83 (and shimming coil signal generator unit 80) nor any transmit signal generator unit 60 and RF sender 63 are added. A trivial solution for a homogeneous field is the zero field. At zero field, the spins remain their initial

orientation. If now a homogeneous field is added, pointing to a different direction, the magnetization will start to perform a precession movement. Hence, an RF pulse was applied. For recording, the main field is ramped to a value, where the MPI recording system is efficient.

[0057] This type of MPI needs some sort of adaption of the sequences. For example, a spin echo is only a reversal of the magnetic field direction. For a volume selection, a field free point (FFP) is moved from infinity to the center of the desired field of view while maintaining a strong gradient. Then the gradient is reduced to zero and the RF sequence is applied.

[0058] There are features in the proposed MPI/MRI combination, which are not possible with classical MRI. For example, the magnetization direction can be turned to a direction where the recording system is most efficient. It is also possible, to bend the B_0 field while recording. This gives more freedom in parallel imaging, as there are three magnetization directions available, in particular for sensitivity encoding (also referred to as SENSE in conventional MRI, described e.g. in K. P. Pruessmann, et al., Sensitivity encoding for fast MRI: SENSE, MAGMA 6(1):87, 1998).

[0059] The possibility to change the main field strength allows the use of contrasts depending on the B_0 dependent T1 e.g. for nitrogen imaging. The scanner is inherently a multi-nuclei system allowing for imaging of hyperpolarized noble gases. Paramagnetic implants have a minor effect on the image quality and scanning is done with minimal heating of implants. So some patients can be scanned with this scanner, who are not allowed to conventional MRI systems. Especially, it is possible to look inside of stents.

[0060] The arbitrary orientation of the B_0 field allows other contrasts, like the unambiguous determination of the tissue electrical conductivity. As B_0 can be low, the conductivity can be imaged for frequencies below 1 MHz, where the tissue variation best reflects pathologies. Moreover frequency dependent conductivities can be recorded.

[0061] Before discussing further specific applications, scenarios for the provision of an "anatomical reference" shall be discussed. MPI images the distribution of magnetic particles. To interpret these data, it is helpful to refer to an image of the underlying anatomy. There are many ways to get this information. The simplest is to register images of other modalities to MPI images using markers that can be identified in both modalities. It is also possible to place a CT system next to an MPI scanner. When using the same patient support for both modalities (called "shared table"), registration of the images becomes simple. Nevertheless, some sort of fusion is still necessary, possibly leading to uncertainties in the fused images. Also, the additional imaging system implies higher cost and more effort in patient handling.

[0062] Therefore, an intrinsic combination of MPI with an anatomical reference image seems to be useful. One such combination would be with magnetic resonance im-

aging which seems to be technically feasible. Both MPI and MRI use strong magnetic fields and record weak oscillating fields. The field geometries and frequencies are not identical, but with moderate technical effort components may be designed serving both MPI and MRI. For example, if the selection field in MPI is generated by electromagnets, a reverse current in one coil yields a homogeneous field suitable for polarizing the protons. The field strength for MRI may reach 0.5 T for resistive coils and 1 T or more for superconducting coils.

[0063] There are a variety of effects that make an intrinsic MPI/MRI combination problematic. For MRI, the fields need to be very homogeneous but simply changing the current direction in one coil usually does not provide a field with sufficient homogeneity because such a coil assembly would be not very efficient for selection field generation. The MPI system contains no sender for the RF B_1 field and no recording system for frequencies as high as several ten MHz. Moreover, the MPI drive and recording coils have self-resonances in the low MHz range making the addition of a MRI B_1 coils difficult.

[0064] One solution for these technical challenges would be to construct a field cycling or pre-polarized MRI system as described by Macovski A, Conolly S, "Novel approaches to low-cost MRI", Magn. Reson. Med. 1993 Aug. 30, (2):221-30. In such a system a strong, but not necessarily very homogeneous field is applied for several hundreds of milliseconds to polarize the protons. After the polarizing pulse the field is reduced to a much lower value where the RF pulses are applied. In MRI, the field homogeneity needs to be roughly $10 \mu\text{T} \mu_0^{-1}$ which is 10 ppm for a 1T scanner but 1% for a 1 mT scanner. Such a value could be reached by changing the current direction in the selection field coils.

[0065] Going to such a low field, the RF pulses can be applied using the drive field coil. The Larmor frequency of protons at 0.6 mT is 25 kHz, i.e. the currently used MPI drive field frequency. The drive field amplitudes in an MPI scanner exceed the necessary amplitudes for MRI by far. An alternative to the drive field for the RF pulse is the focus field. It contains frequency components of up to about 1 kHz. Therefore, the B_0 field has to be even lower, when applying the RF pulse. The focus field amplitude is so high, that the RF pulse may be described as:

- i) Polarize the protons using a strong B_0 field (using the selection field coils, and eventually part of the focus field coils and/or - if available - extra shimming coils).
- ii) Reduce the field strength to zero.
- iii) Increase the field value using a direction which is not identical with the proton polarization (this is the "RF pulse" i.e. the cause of precession). The magnetization direction is preferably rotated for instance by 90° .
- iv) Ramp the field up to achieve a convenient Larmor frequency for signal recording (e.g. 1 MHz).

[0066] A spin echo using this method would be a simple field reversal.

[0067] A pre-polarized MRI would certainly be sufficient for the anatomical reference in MPI. For some selected applications, it may be even superior to the traditional way of MRI. For patients with implants, a pre-polarized MRI would allow near implant imaging as long as the implant is not ferromagnetic. The MPI/MRI combination would be able to operate at even lower frequencies; therefore it would be possible to image inside of metallic stents, possibly assessing re-stenosis. The field cycling allows contrasts, not accessible by conventional MRI. For example, the protein content of tissue can be imaged using the field dependent T_1 relaxation times.

[0068] Advantageously, a MPI and a MRI signal acquisitions can be done in a combined MPI/MRI mode. In particular, while acquiring MPI detection signals at one location in the region of action, the protons can be polarized at other locations (e.g. with a field intensity of up to 0.2 T). The step of polarization is the time-critical step of pre-polarized MRI. Hence, e.g. approximately 10% of the measurement time of MPI can be used for MRI, but nevertheless a MR image having the same quality as a MR image obtained with data acquired during 100% of the time usually required for MRI data acquisition can be obtained. It could also be useful to measure the MR signal at the same location where the MPI signals are obtained since the polarization of the protons will be transported to this location by the blood flow, which enables to obtain a perfusion image.

[0069] As the focus field of an MPI system can point in any direction, a pre-polarized MPI/MRI combination has the ability to direct B_0 arbitrarily. This allows an efficient utilization of the recording coils. Moreover it could solve the problem of current mapping using MRI for electrical impedance measurement in anisotropic tissue. In conventional MRI, only one field component can be measured in the case of low frequency current density imaging. There a phase shift is generated by applying a current through the patient at low frequencies. This is only sensitive to the field component on B_0 -direction. Therefore it is impossible to reconstruct the full current densities. With high frequency current density imaging, two field components can be detected. In this technique, fields due to the currents act as MRI RF fields. Usually, eddy currents in the patient due to B_1 field is utilized. Still, a full reconstruction of the current density is impossible.

[0070] With the proposed prepolarized MPI/MRI combination, all field components can be measured. This allows to reconstruct the anisotropy of the tissue conductivity. Moreover, the ability to choose the Larmor-frequency allows to use current injection frequencies, where minimal effect on the patient is expected but still strong tissue specific contrast is anticipated i.e. at around 100 kHz.

[0071] As such an MPI/MRI system has the ability to rapidly change the field strength and as it is equipped with a very broad-banded receiving system, it may also be well suited for multi-nuclei MRI. As other nuclei than

protons produce relatively weak signal, the presumably low B_0 field strength of an MPI/MRI system is not ideal. On the other hand, there are hyperpolarized techniques in MRI using xenon ^{129}Xe and helium ^3He for lung imaging and carbon ^{13}C for metabolic imaging. Finally, proton lung images with improved quality could be expected, as the susceptibility artefacts are much less severe at low read-out field strength.

[0072] In summary, magnetic particle imaging allows the imaging of fast tracer dynamics, but there is no native tissue contrast. A combination with MRI solves this issue. However, coil geometries in MPI and MRI differ significantly, making direct use impractical. According to one aspect of the present invention it is proposed to use pre-polarized MRI to overcome these difficulties. Further, methods and arrangements are proposed to achieve MRI imaging with minimal additional hardware.

[0073] While the invention has been illustrated and described in detail in the drawings and foregoing description, such illustration and description are to be considered illustrative or exemplary and not restrictive; the invention is not limited to the disclosed embodiments. Other variations to the disclosed embodiments can be understood and effected by those skilled in the art in practicing the claimed invention, from a study of the the appended claims.

Claims

1. An arrangement (10) for influencing and/or detecting magnetic particles in a region of action (300) and for magnetic resonance imaging of an examination object in said region of action (300), which arrangement comprises:

- control means (70) for switching said arrangement between a magnetic particle imaging mode and a magnetic resonance imaging mode,
- a first subset of coils (23) for generating, in the magnetic particle imaging mode, a selection field, namely a stationary magnetic gradient field (211) having a pattern in space of its magnetic field strength such that a first sub-zone (301) having a low magnetic field strength and a second sub-zone (302) having a higher magnetic field strength are formed in the region of action (300),
- a second subset of coils (33) for generating, in the magnetic particle imaging mode, a focus field, namely a temporally changing homogeneous magnetic field for changing the position in space of the region of action (300) and
- a third subset of coils (43) for generating, in the magnetic particle imaging mode, a drive field, namely a temporally changing homogeneous magnetic field for changing the position in space of the two sub-zones (301, 302) in the

region of action (300) so that the magnetization of the magnetic material of the magnetic particles changes locally, the drive field oscillating with a drive field frequency,

wherein said first and/or second subset of coils (23, 33) and/or separate shimming coils (83) are adapted for generating, in the magnetic resonance imaging mode, a substantially homogeneous main magnetic field in a selected and changeable direction, and

wherein said second subset of coils (33) is adapted for generating, in the magnetic resonance imaging mode, a magnetic gradient field in at least two selected and changeable directions,

- an excitation means arranged to excite, in the magnetic resonance imaging mode, magnetic spins in the region of action to precess,

- a receiving means (54) consisting of a receiving coil for selectively acquiring magnetic resonance signals in the magnetic resonance imaging mode and detection signals depending on the magnetization in the region of action (300) in the magnetic particle imaging mode, which magnetization is influenced by the change in the position in space of the first and second sub-zones (301,302),

- signal generation means (20, 30, 40) for generating and providing field generation signals to said subsets of coils, and

- processing means (71) for processing said detection signals and said magnetic resonance signals,

wherein said excitations means comprises:

- a RF signal generator and the receiving coil as a transmit coil of RF pulses, or
- said third subset of coils (43) which is adapted for generating RF transmit signals, or
- said second subset of coils (42) which is adapted for generating RF pulses, or
- means arranged to excite magnetic resonance by cycling the main magnetic field down to a level, where the drive field frequency is the Larmor frequency or by decreasing the main magnetic field down to zero field and increasing the main magnetic field thereafter using a different magnetization direction;

wherein said control means (70) are adapted to control said signal generation means (20, 30, 40) for generating and providing field generation signals to said subsets of coils as required in the respective mode including the magnetic particle imaging mode and the magnetic resonance imaging mode.

2. An arrangement (10) as claimed in claim 1, wherein said first subset of coils (23) comprise a number of at least two, in particular a pair, of first coil elements located on opposite sides of the region of action (300), and
wherein said signal generation means (20) are adapted for providing each first coil element with a separate first field generation signal.
3. An arrangement (10) as claimed in claim 1, wherein said second subset of coils (33) comprises a number of at least six, in particular three pairs (33a, 33b, 33c), of second coil elements located on opposite sides of the region of action, and
wherein said signal generation means (20) are adapted for providing each coil element with a separate second field generation signal.
4. An arrangement (10) as claimed in claim 1, wherein said control means (70) is adapted for controlling said signal generation means, in said magnetic resonance mode, to generate and provide field generation signals to said sub-set of coils to subsequently generate in the region of action
 - a magnetic gradient field by said second subset of coils (33),
 - a first substantially homogenous magnetic field in a first magnetization direction by said first and/or second subset of coils (23, 33),
 - after reducing the field intensity of the first substantially homogenous magnetic field, in particular to substantially zero, a substantially homogenous magnetic field in a second magnetization direction which is substantially different from the first magnetization direction by said first and/or second subset of coils (23, 33).
5. An arrangement (10) as claimed in claim 1, wherein said control means (70) is adapted for controlling said signal generation means, in said magnetic resonance mode, to generate and provide field generation signals to said sub-sets of coils to subsequently
 - generate a strong magnetic gradient field by said second subset of coils (33),
 - reduce the gradient of said magnetic gradient field by said second subset of coils (33) and add a substantially main homogeneous magnetic field by said first and/or second subset of coils (23, 33), and
 - reduce the field strength of said magnetic gradient field by said second subset of coils (33), and

wherein said control means (70) is adapted to control said receiving means (54) to then acquire magnetic

resonance signals.

6. An arrangement (10) as claimed in claim 4 or 5, wherein said control means (70) is adapted for controlling said signal generation means to generate and provide field generation signals to said first and/or third sub-sets of coils and/or to said shimming coils (83) for pre-polarizing protons at portions of the region of action for subsequent acquisition of magnetic resonance signals from said portions, while detection signals are acquired from other portions of said region of action. 5
7. A method for influencing and/or detecting magnetic particles in a region of action (300) and for magnetic resonance imaging of an examination object in said region of action (300), which method comprises the steps of 10
 - generating, in a magnetic particle imaging mode, 20
 - i) a selection field with a first subset of coils, namely a stationary magnetic gradient field (211) having a pattern in space of its magnetic field strength such that a first sub-zone (301) having a low magnetic field strength and a second sub-zone (302) having a higher magnetic field strength are formed in the region of action (300), 25
 - ii) a focus field with a second subset of coils, namely a temporally changing homogeneous magnetic field for changing the position in space of the region of action (300) and 30
 - iii) a drive field with a third subset of coils, namely a temporally changing homogeneous magnetic field for changing the position in space of the two sub-zones (301, 302) in the region of action (300) so that the magnetization of the magnetic material changes locally, the drive field oscillating with a drive field frequency, 35
 - generating, in a magnetic resonance imaging mode, 40
 - iv) by means of said first and/or second subset of coils (23, 33) and/or separate shimming coils (83), a substantially homogeneous main magnetic field in a selected and changeable direction, 45
 - v) by means of said second subset of coils (23, 33), a magnetic gradient field in at least two selected and changeable directions, 50
 - vi) by means of excitation means, RF transmit signals for exciting magnetic resonance in the region of action, 55

- selectively acquiring with a receiving coil magnetic resonance signals in the magnetic resonance imaging mode and detection signals depending on the magnetization in the region of action (300) in the magnetic particle imaging mode, which magnetization is influenced by the change in the position in space of the first and second sub-zones (301, 302), and
- processing said detection signals and said magnetic resonance signals, wherein the RF transmit signals according to step vi) are generated either by means of said receiving coil or by means of coils which are also used for generating said drive field or said focus field, respectively, or in that the main magnetic field is cycled down to a level, where the drive field frequency becomes equal to the Larmor frequency, or in that the main magnetic field is decreased down to zero field and increased thereafter using a different magnetization direction.

8. Computer program comprising program code means for causing a computer to control an arrangement as claimed in claim 1 to carry out the steps of the method as claimed in claim 7 when said computer program is carried out on the computer.

Patentansprüche

1. Anordnung (10) zum Beeinflussen und/oder Detektieren von magnetischen Teilchen in einer Wirkungsregion (300) und zur Magnetresonanzbildgebung eines Untersuchungsobjekts in der genannten Wirkungsregion (300), wobei die Anordnung Folgendes umfasst:
 - Steuerungsmittel (70) zum Umschalten der genannten Anordnung zwischen einem Bildgebungsmodus für magnetische Teilchen und einem Magnetresonanzbildgebungsmodus,
 - eine erste Teilgruppe von Spulen (23), um in dem Bildgebungsmodus für magnetische Teilchen ein Auswahlfeld zu erzeugen, nämlich ein stationäres magnetisches Gradientenfeld (211) mit einem derartigen räumlichen Muster seiner Magnetfeldstärke, dass eine erste Teilzone (301) mit einer niedrigen Magnetfeldstärke und eine zweite Teilzone (302) mit einer höheren Magnetfeldstärke in der Wirkungsregion (300) gebildet werden,
 - eine zweite Teilgruppe von Spulen (33), um in dem Bildgebungsmodus für magnetische Teilchen ein Fokusfeld zu erzeugen, nämlich ein sich zeitlich veränderndes homogenes Magnetfeld zum Verändern der räumlichen Position der Wirkungsregion (300) und
 - eine dritte Teilgruppe von Spulen (43), um in

dem Bildgebungsmodus für magnetische Teilchen ein Ansteuerungsfeld zu erzeugen, nämlich ein sich zeitlich veränderndes homogenes Magnetfeld zum Verändern der räumlichen Position der beiden Teilzonen (301, 302) in der Wirkungsregion (300) auf derartige Weise, dass sich die Magnetisierung des magnetischen Materials der magnetischen Teilchen lokal verändert, wobei das Ansteuerungsfeld mit einer Ansteuerungsfeldfrequenz oszilliert, wobei die genannte erste und/oder zweite Teilgruppe von Spulen (23, 33) und/oder separate Shimming-Spulen (83) dafür ausgelegt sind, im Magnetresonanzbildgebungsmodus ein im Wesentlichen homogenes Hauptmagnetfeld in einer ausgewählten und veränderbaren Richtung zu erzeugen, und wobei die genannte zweite Teilgruppe von Spulen (33) dafür ausgelegt ist, im Magnetresonanzbildgebungsmodus ein magnetisches Gradientenfeld in mindestens zwei ausgewählten und veränderbaren Richtungen zu erzeugen, - ein Anregungsmittel, das dafür vorgesehen ist, im Magnetresonanzbildgebungsmodus magnetische Spins in der Wirkungsregion zum Präzedieren anzuregen, - ein Empfangsmittel (54) bestehend aus einer Empfangsspule zum selektiven Erfassen von Magnetresonanzsignalen im Magnetresonanzbildgebungsmodus und von Detektionssignalen abhängig von der Magnetisierung in der Wirkungsregion (300) im Bildgebungsmodus für magnetische Teilchen, wobei die Magnetisierung durch die Veränderung der räumlichen Position der ersten und der zweiten Teilzone (301, 302) beeinflusst wird, - Signalerzeugungsmittel (20, 30, 40) zum Erzeugen und Bereitstellen von Felderzeugungssignalen für die genannten Teilgruppen von Spulen, und - Verarbeitungsmittel (71) zum Verarbeiten der genannten Detektionssignale und der genannten Magnetresonanzsignale, wobei das genannte Anregungsmittel Folgendes umfasst:

- einen HF-Signalgenerator und die Empfangsspule als eine Sendespule von HF-Impulsen, oder
- die genannte dritte Teilgruppe von Spulen (43), die dafür ausgelegt ist, HF-Sendesignale zu erzeugen, oder
- die genannte zweite Teilgruppe von Spulen (42), die dafür ausgelegt ist, HF-Impulse zu erzeugen, oder
- Mittel, die vorgesehen sind, um Magnetresonanz anzuregen, indem das Hauptmagnetfeld zyklisch bis zu einem Niveau durch-

laufen wird, bei dem die Ansteuerungsfeldfrequenz die Larmor-Frequenz ist, oder indem das Hauptmagnetfeld bis auf null Feld verringert und das Hauptmagnetfeld anschließend unter Verwendung einer anderen Magnetisierungsrichtung erhöht wird;

wobei die genannten Steuermittel (70) dafür ausgelegt sind, die genannten Signalerzeugungsmittel (20, 30, 40) zu steuern, um nach Bedarf in dem jeweiligen Modus, einschließlich des Bildgebungsmodus für magnetische Teilchen und des Magnetresonanzbildgebungsmodus, Felderzeugungssignale für die genannten Teilgruppen von Spulen zu erzeugen und bereitzustellen.

2. Anordnung (10) nach Anspruch 1, wobei die genannte erste Teilgruppe von Spulen (23) eine Anzahl von mindestens zwei, insbesondere ein Paar, von ersten Spulenelementen umfasst, die auf gegenüberliegenden Seiten der Wirkungsregion (300) angeordnet sind, und wobei die genannten Signalerzeugungsmittel (20) dafür ausgelegt sind, für jedes erste Spulenelement ein separates erstes Felderzeugungssignal bereitzustellen.
3. Anordnung (10) nach Anspruch 1, wobei die genannte zweite Teilgruppe von Spulen (33) eine Anzahl von mindestens sechs, insbesondere drei Paare (33a, 33b, 33c), von zweiten Spulenelementen umfasst, die auf gegenüberliegenden Seiten der Wirkungsregion angeordnet sind, und wobei die genannten Signalerzeugungsmittel (20) dafür ausgelegt sind, für jedes Spulenelement ein separates zweites Felderzeugungssignal bereitzustellen.
4. Anordnung (10) nach Anspruch 1, wobei das genannte Steuermittel (70) dafür ausgelegt ist, die genannten Signalerzeugungsmittel in dem genannten Magnetresonanzmodus zu steuern, um Felderzeugungssignale für die genannte Teilgruppe von Spulen zu erzeugen und bereitzustellen, um anschließend in der Wirkungsregion Folgendes zu erzeugen:

- ein magnetisches Gradientenfeld durch die genannte zweite Teilgruppe von Spulen (33),
- ein erstes im Wesentlichen homogenes Magnetfeld in einer ersten Magnetisierungsrichtung durch die genannte erste und/oder zweite Teilgruppe von Spulen (23, 33),
- nach dem Reduzieren der Feldintensität des ersten im Wesentlichen homogenen Magnetfelds, insbesondere auf im Wesentlichen null, ein im Wesentlichen homogenes Magnetfeld in

einer zweiten Magnetisierungsrichtung, die sich im Wesentlichen von der ersten Magnetisierungsrichtung unterscheidet, durch die genannte erste und/oder zweite Teilgruppe von Spulen (23, 33).

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5. Anordnung (10) nach Anspruch 1, wobei das genannte Steuermittel (70) dafür ausgelegt ist, die genannten Signalerzeugungsmittel in dem genannten Magnetresonanzmodus zu steuern, um Felderzeugungssignale für die genannten Teilgruppen von Spulen zu erzeugen und bereitzustellen, um anschließend:

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- ein starkes magnetisches Gradientenfeld durch die genannte zweite Teilgruppe von Spulen (33) zu erzeugen,
- den Gradienten des genannten magnetischen Gradientenfelds durch die genannte zweite Teilgruppe von Spulen (33) zu reduzieren und ein im Wesentlichen homogenes Hauptmagnetfeld durch die genannte erste und/oder zweite Teilgruppe von Spulen (23, 33) hinzuzufügen, und
- die Feldstärke des genannten magnetischen Gradientenfelds durch die genannte zweite Teilgruppe von Spulen (33) zu reduzieren, und

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wobei das genannte Steuermittel (70) dafür ausgelegt ist, die genannten Empfangsmittel (54) zu steuern, um dann Magnetresonanzsignale zu erfassen.

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6. Anordnung (10) nach Anspruch 4 oder 5, wobei das genannte Steuermittel (70) dafür ausgelegt ist, die genannten Signalerzeugungsmittel zu steuern, um Felderzeugungssignale für die genannte erste und/oder dritte Teilgruppe von Spulen und/oder die genannten Shimming-Spulen (83) zu erzeugen und bereitzustellen, um Protonen bei Bereichen der Wirkungsregion für die anschließende Erfassung von Magnetresonanzsignalen aus den genannten Bereichen vorzupolarisieren, während Detektionssignale aus anderen Bereichen der genannten Wirkungsregion erfasst werden.

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7. Verfahren zum Beeinflussen und/oder Detektieren von magnetischen Teilchen in einer Wirkungsregion (300) und zur Magnetresonanzbildgebung eines Untersuchungsobjekts in der genannten Wirkungsregion (300), wobei das Verfahren Folgendes umfasst:

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- Erzeugen, in einem Bildgebungsmodus für magnetische Teilchen,

- i) eines Auswahlfelds mit einer ersten Teilgruppe von Spulen, nämlich eines stationären magnetischen Gradientenfelds (211) mit einem derartigen räumlichen Mus-

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ter seiner Magnetfeldstärke, dass eine erste Teilzone (301) mit einer niedrigen Magnetfeldstärke und eine zweite Teilzone (302) mit einer höheren Magnetfeldstärke in der Wirkungsregion (300) gebildet werden, ii) eines Fokusfelds mit einer zweiten Teilgruppe von Spulen, nämlich eines sich zeitlich verändernden homogenen Magnetfelds zum Verändern der räumlichen Position der Wirkungsregion (300) und iii) eines Ansteuerungsfelds mit einer dritten Teilgruppe von Spulen, nämlich eines sich zeitlich verändernden homogenen Magnetfelds zum Verändern der räumlichen Position der beiden Teilzonen (301, 302) in der Wirkungsregion (300) auf derartige Weise, dass sich die Magnetisierung des magnetischen Materials lokal verändert, wobei das Ansteuerungsfeld mit einer Ansteuerungsfeldfrequenz oszilliert,

- Erzeugen, in einem Magnetresonanzbildgebungsmodus,

- iv) mithilfe der genannten ersten und/oder zweiten Teilgruppe von Spulen (23, 33) und/oder separater Shimming-Spulen (83), eines im Wesentlichen homogenen Hauptmagnetfelds in einer ausgewählten und veränderbaren Richtung,
- v) mithilfe der genannten zweiten Teilgruppe von Spulen (33), eines magnetischen Gradientenfelds in mindestens zwei ausgewählten und veränderbaren Richtungen,
- vi) mithilfe eines Anregungsmittels, von HF-Sendesignalen zum Anregen von Magnetresonanz in der Wirkungsregion,

- selektives Erfassen von Magnetresonanzsignalen im Magnetresonanzbildgebungsmodus und von Detektionssignalen abhängig von der Magnetisierung in der Wirkungsregion (300) im Bildgebungsmodus für magnetische Teilchen mit einer Empfangsspule, wobei die Magnetisierung durch die Veränderung der räumlichen Position der ersten und der zweiten Teilzone (301, 302) beeinflusst wird, und

- Verarbeiten der genannten Detektionssignale und der genannten Magnetresonanzsignale, wobei die HF-Sendesignale nach Schritt vi) entweder mithilfe der genannten Empfangsspule oder mithilfe von Spulen erzeugt werden, die auch zum Erzeugen des genannten Ansteuerungsfelds bzw. des genannten Fokusfelds verwendet werden, oder indem das Hauptmagnetfeld zyklisch bis zu einem Niveau durchlaufen wird, bei dem die Ansteuerungsfeldfrequenz gleich der Larmor-Frequenz wird, oder indem

das Hauptmagnetfeld bis auf null Feld verringert wird und anschließend unter Verwendung einer anderen Magnetisierungsrichtung erhöht wird.

8. Computerprogramm umfassend Programmcode-
mittel zum Veranlassen eines Computers, eine An-
ordnung nach Anspruch 1 zu steuern, um die Schritte
des Verfahrens nach Anspruch 7 durchzuführen,
wenn das genannte Computerprogramm auf dem
Computer ausgeführt wird.

Revendications

1. Agencement (10) pour influencer et/ou détecter des
particules magnétiques dans une région d'action
(300) et pour une imagerie par résonance magnéti-
que d'un objet d'examen dans ladite région d'action
(300), ledit agencement comprenant :
- des moyens de commande (70) pour faire al-
terner ledit agencement entre un mode d'ima-
gerie de particules magnétiques et un mode
d'imagerie par résonance magnétique,
 - un premier sous-ensemble de bobines (23)
pour générer, dans le mode d'imagerie de par-
ticules magnétiques, un champ de sélection, à
savoir un champ de gradient magnétique sta-
tionnaire (211) ayant un motif dans l'espace de
son intensité de champ magnétique tel qu'une
première sous-zone (301) ayant une faible in-
tensité de champ magnétique et une seconde
sous-zone (302) ayant une plus forte intensité
de champ magnétique sont formées dans la ré-
gion d'action (300),
 - un deuxième sous-ensemble de bobines (33)
pour générer, dans le mode d'imagerie de par-
ticules magnétiques, un champ de focalisation,
à savoir un champ magnétique homogène va-
riable dans le temps pour modifier la position
dans l'espace de la région d'action (300) et
 - un troisième sous-ensemble de bobines (43)
pour générer, dans le mode d'imagerie de par-
ticules magnétiques, un champ d'entraînement,
à savoir un champ magnétique homogène va-
riable dans le temps pour modifier la position
dans l'espace des deux sous-zones (301, 302)
dans la région d'action (300) de sorte que la ma-
gnétisation de la matière magnétique des parti-
cules magnétiques change localement, le
champ d'entraînement oscillant avec une fré-
quence de champ d'entraînement,
dans lequel lesdits premier et/ou deuxième
sous-ensembles de bobines (23, 33) et/ou des
bobines de compensation séparées (83) sont
adaptés pour générer, dans le mode d'imagerie
par résonance magnétique, un champ magné-
tique principal sensiblement homogène dans

une direction sélectionnée et modifiable, et
dans lequel ledit deuxième sous-ensemble de
bobines (33) est adapté pour générer, dans le
mode d'imagerie par résonance magnétique, un
champ de gradient magnétique dans au moins
deux directions sélectionnées et modifiables,
- un moyen d'excitation agencé pour exciter,
dans le mode d'imagerie par résonance magné-
tique, des spins magnétiques dans la région
d'action à précessionner,
- un moyen de réception (54) constitué d'une
bobine réceptrice pour sélectivement acquérir
des signaux de résonance magnétique dans le
mode d'imagerie par résonance magnétique et
des signaux de détection en fonction de la ma-
gnétisation dans la région d'action (300) dans
le mode d'imagerie de particules magnétiques,
ladite magnétisation étant influencée par le
changement de position dans l'espace des pre-
mière et seconde sous-zones (301, 302),
- des moyens de génération de signaux (20, 30,
40) pour générer et fournir des signaux de gé-
nération de champ auxdits sous-ensembles de
bobines, et
- des moyens de traitement (71) pour traiter les-
dits signaux de détection et lesdits signaux de
résonance magnétique,
dans lequel ledit moyen d'excitation comprend :

- un générateur de signaux RF et la bobine
réceptrice comme bobine de transmission
d'impulsions RF, ou
- ledit troisième sous-ensemble de bobines
(43) qui est adapté pour générer des si-
gnaux de transmission RF, ou
- ledit deuxième sous-ensemble de bobines
(42) qui est adapté pour générer des impul-
sions RF, ou
- des moyens agencés pour exciter une ré-
sonance magnétique en abaissant le
champ magnétique principal à un niveau où
la fréquence de champ d'entraînement est
la fréquence de Larmor ou en diminuant le
champ magnétique principal jusqu'à un
champ nul et en augmentant ensuite le
champ magnétique principal en utilisant
une direction de magnétisation différente ;

dans lequel lesdits moyens de commande (70)
sont adaptés pour commander auxdits moyens
de génération de signaux (20, 30, 40) de générer
et fournir des signaux de génération de champ
auxdits sous-ensembles de bobines comme ce-
la est exigé dans le mode respectif comprenant
le mode d'imagerie de particules magnétiques
et le mode d'imagerie par résonance magnéti-
que.

2. Agencement (10) selon la revendication 1, dans lequel ledit premier sous-ensemble de bobines (23) comprennent un nombre d'au moins deux, en particulier une paire, de premiers éléments bobines situés sur des côtés opposés de la région d'action (300), et dans lequel lesdits moyens de génération de signaux (20) sont adaptés pour fournir à chaque premier élément bobine un premier signal séparé de génération de champ.
3. Agencement (10) selon la revendication 1, dans lequel ledit deuxième sous-ensemble de bobines (33) comprend un nombre d'au moins six, en particulier trois paires (33a, 33b, 33c), de deuxième éléments bobines situés sur des côtés opposés de la région d'action, et dans lequel lesdits moyens de génération de signaux (20) sont adaptés pour fournir à chaque élément bobine un deuxième signal séparé de génération de champ.
4. Agencement (10) selon la revendication 1, dans lequel lesdits moyens de commande (70) sont adaptés pour commander auxdits moyens de génération de signaux, dans ledit mode de résonance magnétique, de générer et fournir des signaux de génération de champ audit sous-ensemble de bobines pour faire générer ensuite dans la région d'action
- un champ de gradient magnétique par ledit deuxième sous-ensemble de bobines (33),
 - un premier champ magnétique sensiblement homogène dans une première direction de magnétisation par lesdits premier et/ou deuxième sous-ensembles de bobines (23, 33),
 - après réduction de l'intensité de champ du premier champ magnétique sensiblement homogène, en particulier sensiblement à zéro, un champ magnétique sensiblement homogène dans une deuxième direction de magnétisation qui est sensiblement différente de la première direction de magnétisation par lesdits premier et/ou deuxième sous-ensembles de bobines (23, 33).
5. Agencement (10) selon la revendication 1, dans lequel lesdits moyens de commande (70) sont adaptés pour commander auxdits moyens de génération de signaux, dans ledit mode de résonance magnétique, de générer et fournir des signaux de génération de champ auxdits sous-ensembles de bobines pour ensuite
- faire générer un champ de fort gradient magnétique par ledit deuxième sous-ensemble de bobines (33),
 - faire réduire le gradient dudit champ de gradient magnétique par ledit deuxième sous-ensemble de bobines (33) et faire ajouter un champ magnétique principal sensiblement homogène par lesdits premier et/ou deuxième sous-ensembles de bobines (23, 33), et
 - faire réduire l'intensité de champ dudit champ de gradient magnétique par ledit deuxième sous-ensemble de bobines (33), et
- dans lequel lesdits moyens de commande (70) sont adaptés pour commander audit moyen de réception (54) d'acquiesir ensuite des signaux de résonance magnétique.
6. Agencement (10) selon les revendications 4 ou 5, dans lequel lesdits moyens de commande (70) sont adaptés pour commander auxdits moyens de génération de signaux de générer et fournir des signaux de génération de champ auxdits premier et/ou troisième sous-ensembles de bobines et/ou auxdites bobines de compensation (83) pour pré-polariser des protons dans des parties de la région d'action pour une acquisition ultérieure de signaux de résonance magnétique provenant desdites parties, pendant que des signaux de détection sont acquis à partir d'autres parties de ladite région d'action.
7. Procédé pour influencer et/ou détecter des particules magnétiques dans une région d'action (300) et pour une imagerie par résonance magnétique d'un objet d'examen dans ladite région d'action (300), ledit procédé comprenant les étapes qui consistent à :
- générer, dans un mode d'imagerie de particules magnétiques,
 - i) un champ de sélection avec un premier sous-ensemble de bobines, à savoir un champ de gradient magnétique stationnaire (211) ayant un motif dans l'espace de son intensité de champ magnétique tel qu'une première sous-zone (301) ayant une faible intensité de champ magnétique et une seconde sous-zone (302) ayant une plus forte intensité de champ magnétique sont formées dans la région d'action (300),
 - ii) un champ de focalisation avec un deuxième sous-ensemble de bobines, à savoir un champ magnétique homogène variable dans le temps pour modifier la position dans l'espace de la région d'action (300) et
 - iii) un champ d'entraînement avec un troisième sous-ensemble de bobines, à savoir un champ magnétique homogène variable dans le temps pour modifier la position dans l'espace des deux sous-zones (301, 302) dans la région d'action (300) de sorte que la magnétisation de la matière magnétique change localement, le champ d'entraîne-

ment oscillant avec une fréquence de champ d'entraînement,

- générer, dans un mode d'imagerie par résonance magnétique, 5

iv) au moyen desdits premier et/ou deuxième sous-ensembles de bobines (23, 33) et/ou desdites bobines de compensation séparées (83), un champ magnétique principal sensiblement homogène dans une direction sélectionnée et modifiable, 10

v) au moyen dudit deuxième sous-ensemble de bobines (33), un champ de gradient magnétique dans au moins deux directions sélectionnées et modifiables, 15

vi) au moyen de moyens d'excitation, des signaux de transmission RF pour exciter une résonance magnétique dans la région d'action, 20

- acquérir sélectivement avec une bobine réceptrice des signaux de résonance magnétique dans le mode d'imagerie par résonance magnétique et des signaux de détection en fonction de la magnétisation dans la région d'action (300) dans le mode d'imagerie de particules magnétiques, ladite magnétisation étant influencée par le changement de position dans l'espace des première et seconde sous-zones (301, 302), et 25 30

- traiter lesdits signaux de détection et lesdits signaux de résonance magnétique, dans lequel les signaux de transmission RF selon l'étape vi) sont générés soit au moyen de ladite bobine réceptrice, soit au moyen de bobines qui servent également à générer ledit champ d'entraînement ou ledit champ de focalisation, respectivement, ou dans lequel le champ magnétique principal est abaissé à un niveau où la fréquence de champ d'entraînement devient égale à la fréquence de Larmor, ou dans lequel le champ magnétique principal est réduit à un champ nul et augmenté ensuite à l'aide d'une direction de magnétisation différente. 35 40 45

8. Programme informatique comprenant des moyens de code de programme pour amener un ordinateur à commander à un agencement selon la revendication 1 d'effectuer les étapes du procédé selon la revendication 7 lorsque ledit programme informatique est exécuté sur l'ordinateur. 50

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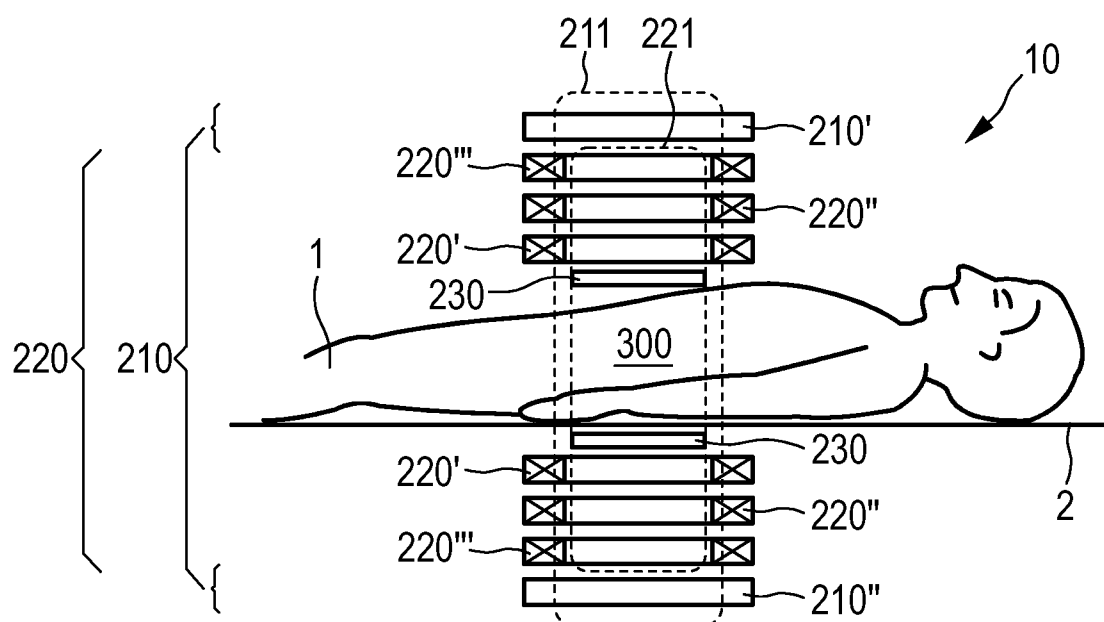


FIG. 1

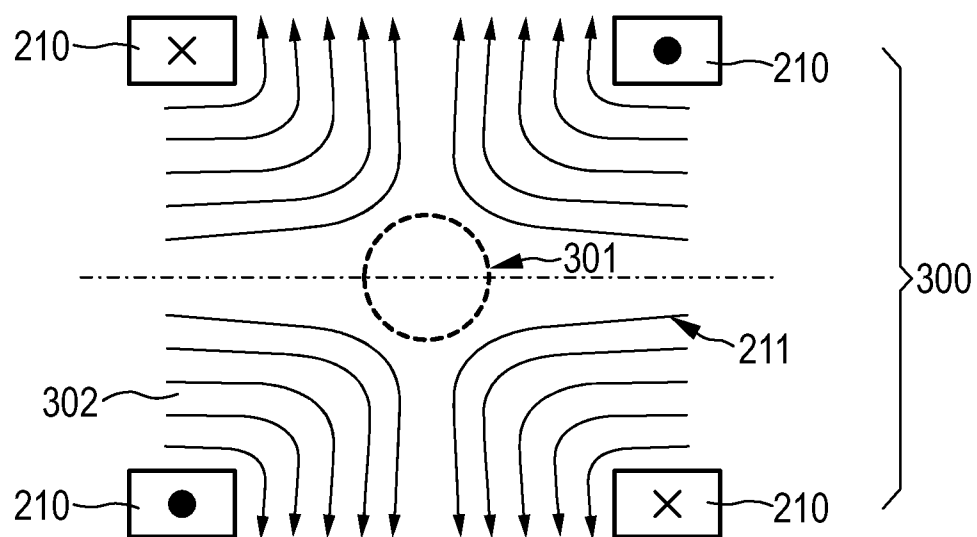


FIG. 2

FIG. 3

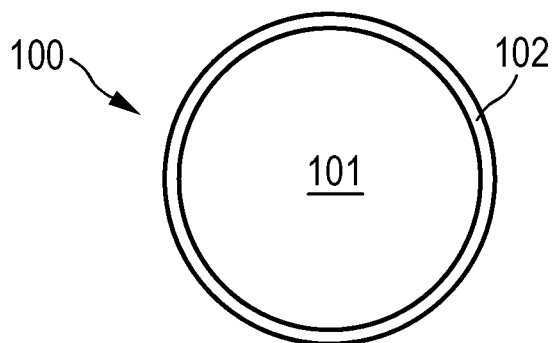


FIG. 4a

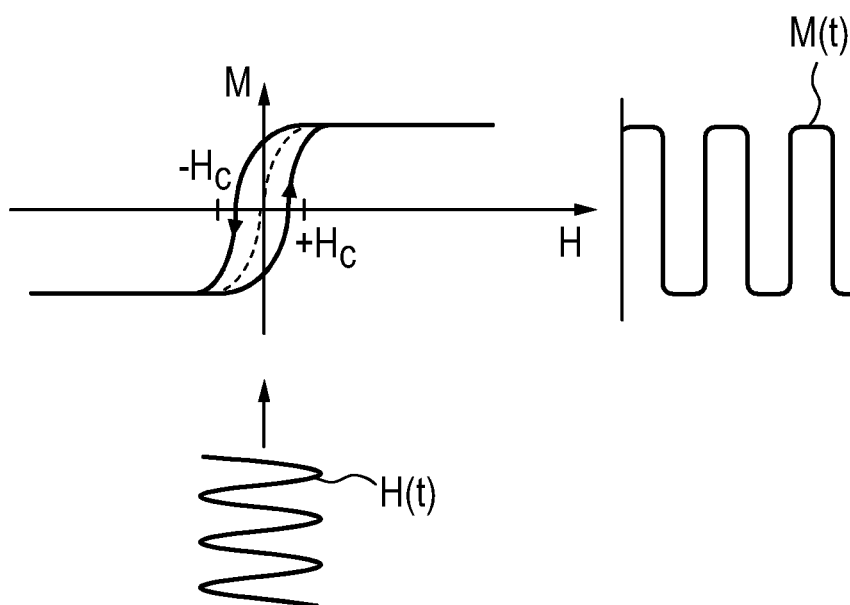
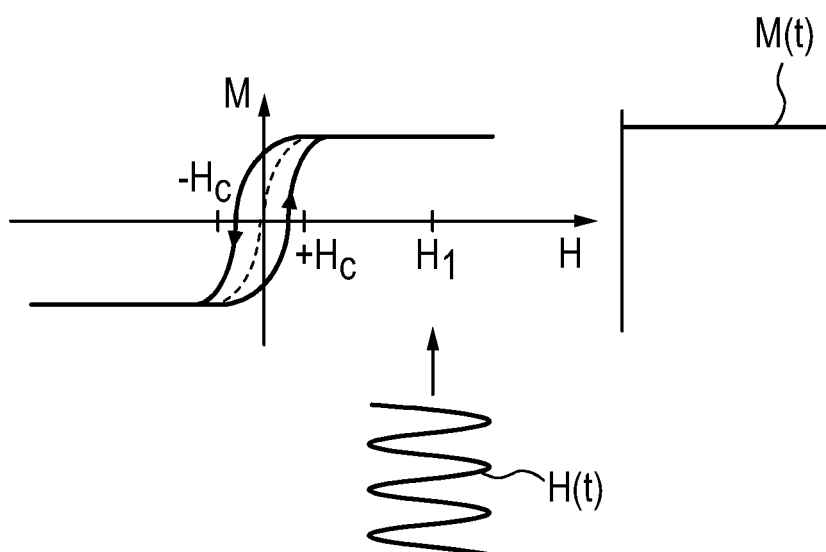


FIG. 4b



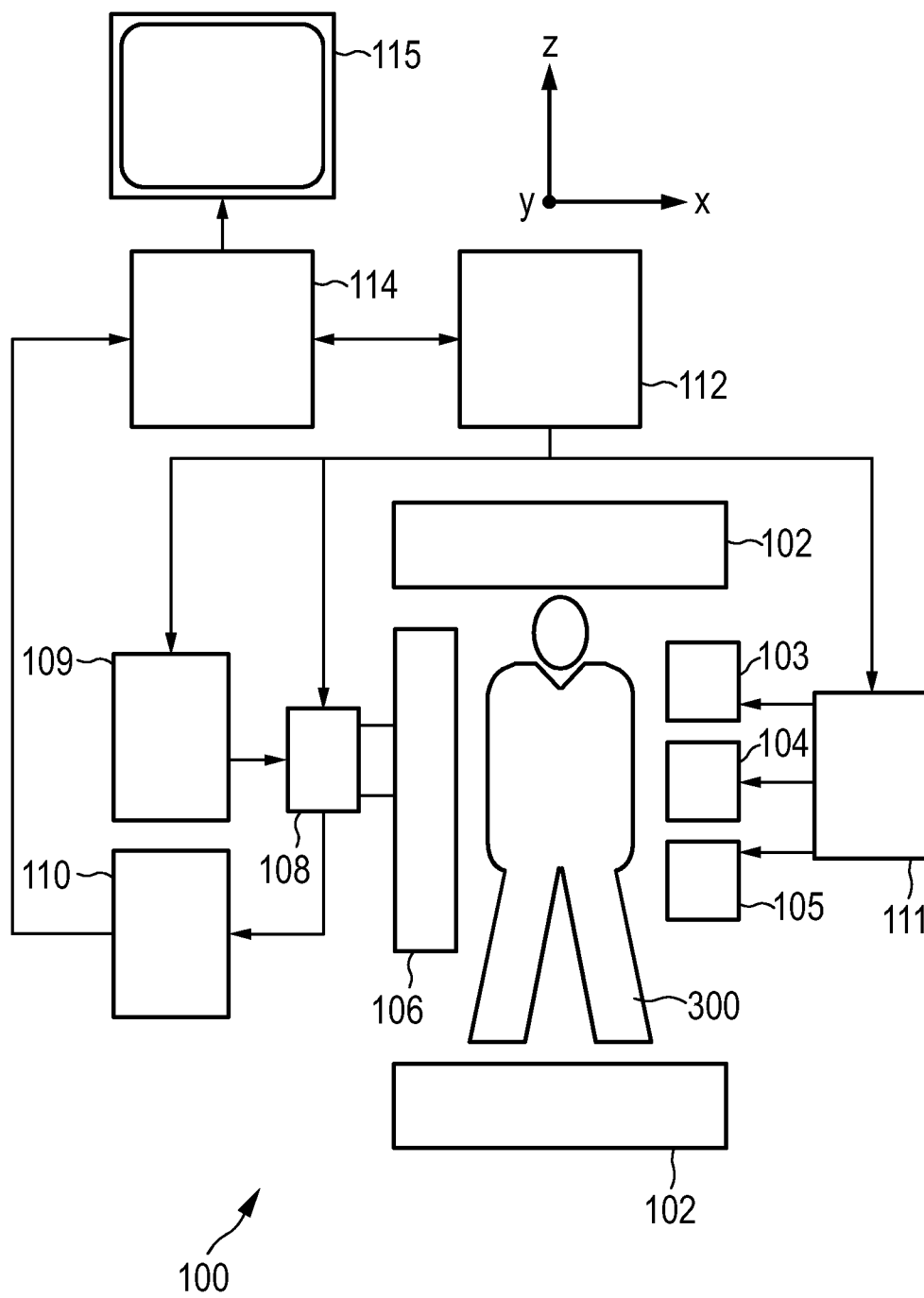


FIG. 5

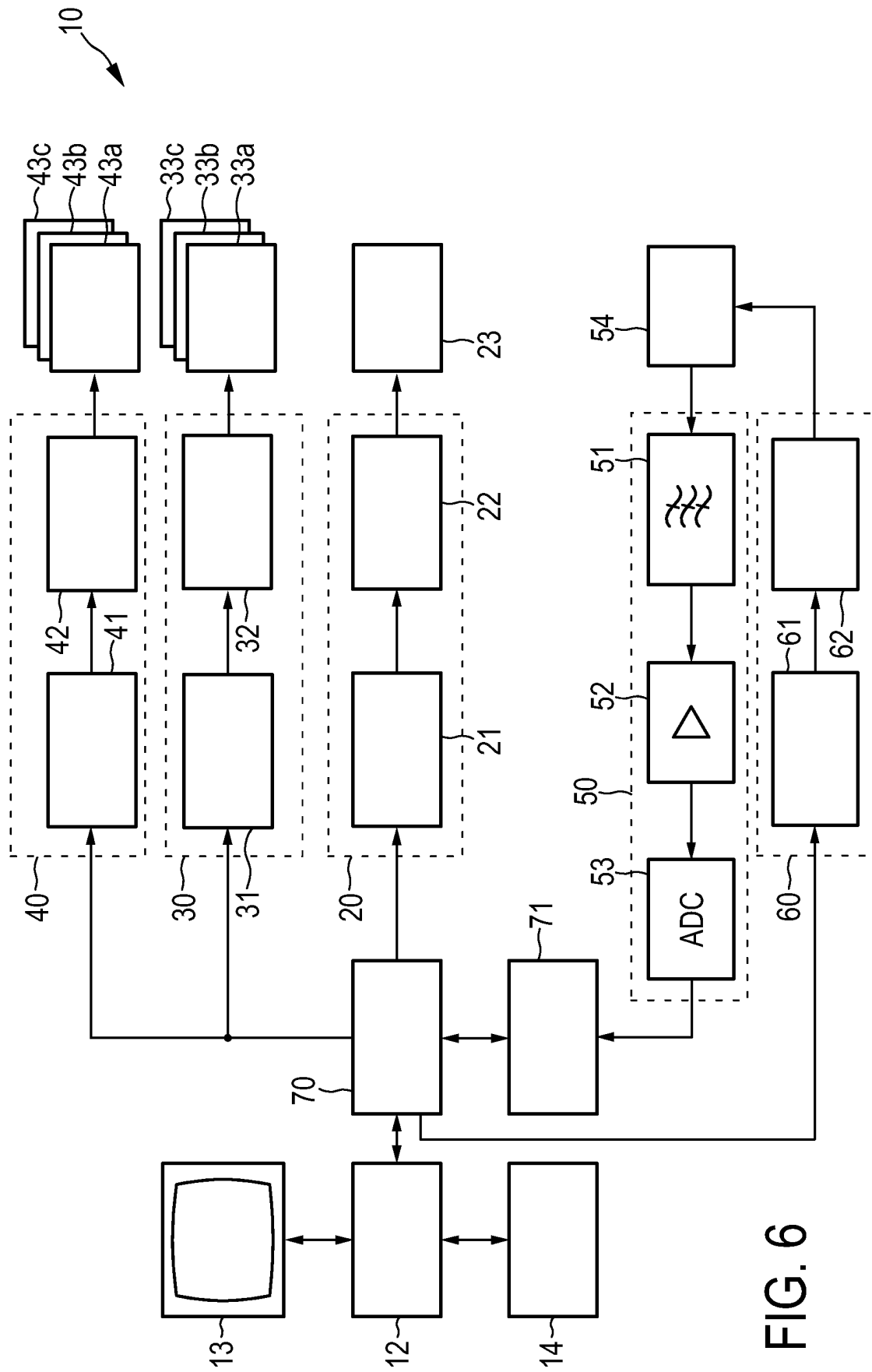


FIG. 6

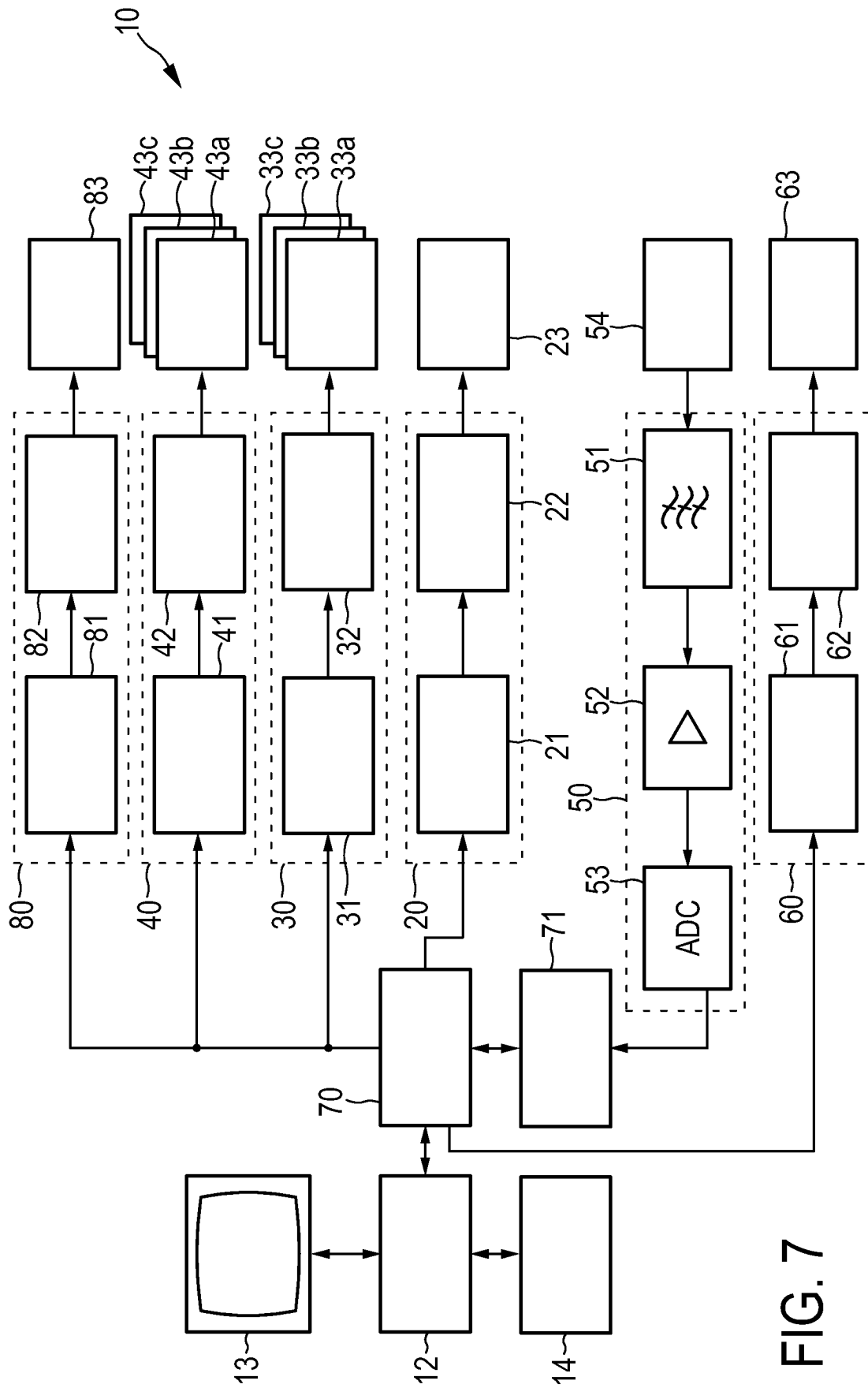


FIG. 7

REFERENCES CITED IN THE DESCRIPTION

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专利名称(译)	用于影响和/或检测磁性粒子和用于磁共振成像的装置和方法		
公开(公告)号	EP2425268B1	公开(公告)日	2017-08-02
申请号	EP2010717844	申请日	2010-04-23
[标]申请(专利权)人(译)	皇家飞利浦电子股份有限公司		
申请(专利权)人(译)	皇家飞利浦电子N.V. 飞利浦知识产权及标准部GMBH		
当前申请(专利权)人(译)	皇家飞利浦N.V. 飞利浦知识产权及标准部GMBH		
[标]发明人	GLEICH BERNHARD		
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CPC分类号	A61B5/0035 A61B5/05 A61B5/0515 G01R33/381 G01R33/3875 G01R33/445 G01R33/4808		
优先权	2009159257 2009-04-30 EP		
其他公开文献	EP2425268A1		
外部链接	Espacenet		

摘要(译)

磁性粒子成像允许快速示踪动力学的成像，但没有天然组织对比度。与MRI的组合解决了这个问题。然而，MPI和MRI中的线圈几何形状显着不同，使得直接使用不切实际。根据本发明的一个方面，提出使用预极化MRI来克服这些困难。此外，提出了用最少的附加硬件实现MRI成像的方法和装置。

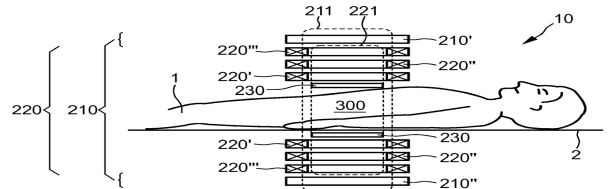


FIG. 1

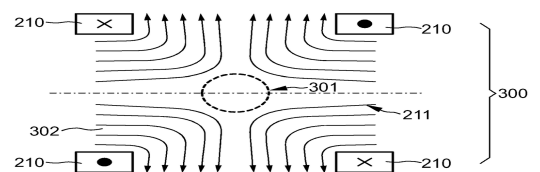


FIG. 2