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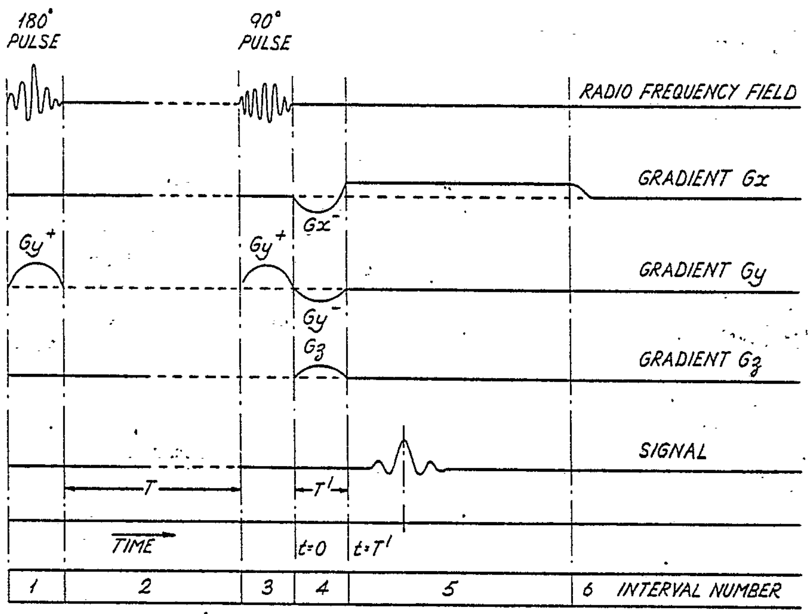

(57) Abstract

To produce image information from an object it is subjected to a continuous static magnitic field along a Z axis and to sets of sequences of orthogonal gradients G_X, G_Y and G_Z to the magnetic field. Spins in a selected plane (the X-Z plane) are excited by selective rf pulses and an associated G_Y gradient and the selected spins are subjected to all three gradients of which the G_Z gradient provides twist or warp to each column of spins extending along the Z axis to phase-encode the columns. The spin-echo signals are read out in the presence of a G_X gradient. In each set of sequences a different value of Z gradient is employed. The Fourier transformed spin-echo signals obtained from each sequence, when arranged in order of increasing G_Z gradient and subjected to a second Fourier transform represent the distribution of spin density in the Z direction, thus giving a two-dimensional image of the selected X-Z plane.

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METHODS OF PRODUCING IMAGE INFORMATION FROM OBJECTS

The present invention relates to methods of producing image information from objects. It is concerned with producing images of samples containing nuclear or other spins whose spatial distribution of density or relaxation time is detected by magnetic resonance techniques. More particularly, it describes methods for producing images from free induction decays (FID's) and spin echoes of the sample in the presence of static magnetic fields and switched magnetic field gradients.

In U.S. Patent No. 4,070,611 there is described a method of producing images by a series of FID's following separate excitations of the sample. During these FID's, magnetic fields in two (or three) orthogonal directions are switched on and off for specific lengths of time to yield two (or three) dimensional images.

One of the problems associated with the above method is that inhomogeneities of the static magnetic field can simulate the effect of the deliberately introduced switched filed gradients and mask the effect of these switched field gradients, thereby destroying some of the information contained in the signal.

The masking effect can occur as follows. Different FID's have field gradients of fixed strength switched on for varying times. For any particular combination of gradient pulse lengths during a single FID, spins in different regions of the sample experience varying phase shifts relative to each other.

These phase shifts allow spatial discrimination and therefore enable an image to be formed. The amount of phase shift between two regions of the sample is proportional to the difference in local magnetic field of the two regions. If inhomogeneities in the static magnetic field contribute to the local field (in addition to deliberately introduced gradients), the spatial distribution information can be distorted.

Suppose for example that in one of the FID's needed to produce an image, the gradient G_z in the z direction is switched on for time T. Consider a small volume element of a sample at

$Z = Z_o$. It experiences a field $B(Z) = B_o + Z_oG_z + \Delta B(Z_o)$, where B_o is the static magnetic field at $Z = 0$ and $\Delta B(Z_o)$ is the deviation from the static field at Z_o due to the inhomogeneity in the static field. Then at the end of time T the spins at $Z = Z_o$ experience a change in phase $\Delta \emptyset$ relative to spins at $Z = 0$ given by

$$\Delta\emptyset = \gamma(Z_oG_z + \Delta B(Z_o))T \qquad (1)$$

where γ is the gyromagnetic ratio. So effectively the gradient appears to be not G_z but $G_z + \Delta B(Z_o)/Z_o$. A Fourier transform along the Z direction will yield a distorted and non-linear scale in that direction. Moreover, the extra phase shifts could cause signals from some parts of the sample to appear incorrectly in the wrong part of the image (aliasing).

A numerical example will illustrate the seriousness of this problem.

In order to produce a well-determined NxN image, it is necessary to take N samples from N signals. For whole-body imaging, one would require a region at least 40 cm in diameter and produce a 64 x 64 element image.

Referring to equation (1), there need to be 64 different values of $\Delta\emptyset$. These are obtained by having 64 different values of time during which the gradient G_z is applied. Ignoring the ΔB term in equation (1) for the moment, an example of such a series is given by

$$\Delta\emptyset_1 = 0$$
$$\Delta\emptyset_2 = \Delta\emptyset^* = (\gamma Z_oG_z)\ (T)$$
$$\Delta\emptyset_3 = 2\Delta\emptyset^* = (\gamma Z_oG_z)\ (2T)$$
$$\Delta\emptyset_4 = 3\Delta\emptyset^* = (\gamma Z_oG_z)\ (3T)$$
$$\Delta\emptyset_N = (N - 1)\Delta\emptyset^* = (\gamma Z_oG_z)\ \left[(N - 1)\ T\right]$$

Note that the stength of the gradient is the same for each $\Delta\emptyset_k$ but the time the gradients are applied varies. There is a condition, however, that $\Delta\emptyset^* < 2\Pi$ across the sample. Using reasonable parameters based on a whole-body nuclear magnetic resonance imaging reference, the length of the sample can be set to L = 40 cm, T = 0.5 ms, $\gamma/2\Pi$ = 4260 Hz/Gauss, and the condition $\Delta\phi^* < 2\,\Pi$ gives $G_z < 0.012$ Gauss/cm. At 20 cm, the maximum distance from the centre of the field, $G_z \times Z = .24$ Gauss. But the inhomogeneity in a four coil, eighth order resistive magnet (which is typical of those used for whole-body imaging) will be about 10^{-4} at 20 cm, or .1 Gauss for a 1 kGauss magnet, nearly half the contribution of the gradient. This situation is unacceptable since one is trying to resolve those 20 cm into 32 parts, and the distortion introduced by such inhomogeneity would ruin the image making process.

Looked at in a slightly different way, the above difficulty imposes a stringent condition on the homogeneity of the static magnetic field.

The principal object of the present invention is to provide an improved method of gyromagnetic resonance imaging employing trains of free induction decays.

According to the present invention a method of deriving image information from an object using nuclear magnetic resonance signals comprises subjecting an object to a continuous static magnetic field along one axis and repeatedly carrying out a set of sequential steps each set of steps comprising:

1) selectively exciting nuclear spins in a plane in the presence of a first gradient of the magnetic field which has a gradient direction perpendicular to said plane;

2) reversing the direction of the said first gradient and applying a second gradient and a third gradient of the magnetic field, the direction of the second gradient being orthogonal to the gradient direction of the first gradient, and the third gradient having a gradient direction orthogonal to the gradient directions of both the first and second gradients;

3) reversing the direction of the second gradient of the magnetic field and holding the said reversed gradient while reading out the resultant free induction decay signal from the said object;

and then successively repeating the above set of steps with different values of gradient of the third gradient, there being a recovery interval between the repetition of successive sets of steps.

Preferably the period of application of the said third gradient of the magnetic field is equal in each of the sets of steps.

In carrying out the invention it may be convenient to apply the first, second and third gradients of the magnetic field in step 2) simultaneously.

In order that the invention may be more fully understood reference will now be made to the accompanying drawings in which the single figure shows pulse sequences for an embodiment of the invention.

For the purpose of the following description a static magnetic field B_o lies along the Z axis and the radiofrequency (rf) field lies along the Y axis. There are coils to produce gradients G_x, G_y and G_z to the magnetic field B_o in the X,Y and Z directions. The production of a two-dimensional image of a thin slab perpendicular to the Y axis is considered. Where the method is applied to human whole body imaging it is convenient to position the patient horizontally with the Z direction vertical and the Y direction horizontal along the length of the patient. The X direction is then horizontal across the patient.

The pulse sequence used to form an image from single spin echoes following separate excitations is shown in Figure 1. The time axis is divided into six into six successive intervals repeated cyclically. The fields which are applied in each of these intervals are as follows:

Interval 1. A 180^o rf pulse is applied simultaneously with a magnetic field gradient G_y^+. This selectively inverts the nuclear spins in and close to the plane $Y=Y_o$. The value of Y_o can be altered by a change in the frequency of the 180^o pulse. Alternatively, a non-selective 180^o pulse can be applied with no gradients present. Yet again non-selective spin inversion can be obtained by an adiabatic fast passage in which the rf field is swept through a frequency range. y-dimension selection is then done entirely by the events in interval 3.

Interval 2. The nuclear spin system is allowed to relax by spin-lattice relaxation for a chosen time T. No fields other than B_o are applied during this interval.

Interval 3. A weak 90^o rf pulse is applied simultaneously with a magnetic field gradient G_y^+. This selectively excites nuclear spins in and close to the plane $Y=Y_o$. The value of Y_o can be altered by a change in the frequency of the 90^o pulse.

Interval 4. A gradient of the magnetic field having a negative value G_y^- is applied to rephase the selected nuclear spins along the Y direction. Simultaneously a negative gradient G_x^- to the magnetic field is applied to dephase the nuclear spins along the X direction. Simultaneously a gradient G_z of the magnetic field is applied to dephase the spins along Z.

Interval 5. A smaller positive magnetic field gradient G_x^+ is applied. During this interval, the nuclear spins rephase to form a spin echo, when the free induction signal is a maximum, and then dephase. It is desirable to keep G_x^+ constant during this interval, at which time the nuclear free induction signal is collected.

Interval 6. System recovery time until the occurrence of interval 1 of the next sequence. This should be long compared with the spin-lattice relaxation time T and is of the order of a second in the whole-body imaging apparatus described in reference 4.

The various gradients of the magnetic field described above need not have a square wave time profile but can have a sinusoidal profile of amplitude against time, which makes less demand on the switching circuits for the gradient coils.

Two different kinds of free induction signals, S_A and S_B are obtained using this pulse sequence, for any one value of the altered high frequency of interval 1 or interval 3 and any one value of $\int_4 dt\ G_z$ in interval 4, where $\int_4 dt$ indicates the integral over interval 4.

S_B: The relaxation interval (2) is comparable with the spin-lattice relaxation time being measured. That is, $T \approx T_1$, which is a few hundred milliseconds for human soft tissue at 1.7 MHz.

S_A: The events of intervals 1 and 2 are omitted, but the rest of the sequence is identical.

S_A contains mainly proton density information, and S_B contains both spin-lattice relaxation time (T_1) information and proton density information.

It may be desirable to consider the events in interval 4 in more detail. In that interval all three orthogonal magnetic field gradients are applied simultaneously. At first sight this would appear to make analysis of the spin behaviour rather difficult but since no radio frequency field is present in interval 4 the effects of the three gradients can be considered separately. The resultant effect of the three gradients although applied simultaneously is the same as if they were applied sequentially and the coincidence in timing is simply a convenience which serves to save time between excitation and signal acquisition in interval 5. The G_y^- gradient serves to rephase the spins across the width of the selected slice and thus it maximises the signal that is eventually obtained. The G_x^- gradient serves to dephase the spins along the X direction as a preliminary to the readout step in interval 5 in which a G_x^+ gradient is applied to cause the spins to rephase and produce a kind of gradient-induced spin echo in the middle of the signal acquisition period. The Fourier transform of this spin echo signal is therefore a one-dimensional projection of the spin density within the slice on to the X axis.

The function of the gradient G_z is to provide discrimination in the Z direction. It is introduced in interval 4 to give a known amount of twist or "warp" to each vertical column of spins (Z axis vertical), and thus it phase encodes the signal prior to projection on to the X axis. In fact it maximises the response to a particular vertical spatial frequency in the column equal to the spatial frequency of the "warp".

The entire set of steps described above is successively repeated a number of times and in each repetition a different amplitude of gradient G_z is utilised to cover a range of vertical spatial frequencies from zero up to a maximum. Now suppose that the projected spin density values for any one column (obtained from the Fourier transforms of the spin-echo signals) are arranged in order of increasing G_z pulse size, and subjected to another Fourier transform, then this will represent the distribution of spin density up the column. When this is done for each column a complete two-dimensional image of the selected slice is obtained. Thus, an N x N image can be obtained by taking N projections on to the <u>same</u> axis, a procedure which is obviously impossible in X-ray or radio-isotope imaging, simply because the basic signal has no phase information in it.

Another view of the action of the phase-encoding gradient G_z is that in each imaging column N projections are collected onto the X axis. The projections are different because spins at different heights are given varying amounts of phase twist by the presence of different values of G_z (hence the name "spin warp").

The phase information in an NMR signal is preserved by employing two phase-sensitive detectors in quadrature to produce two signals which are then treated as a single complex number. The outcome of the double Fourier transformation is a matrix of complex numbers whose amplitudes represent the required spin densities. Their phases would ideally be identical in a perfect magnetic field; in practice they may vary considerably by many cycles over the image plane, representing as they do the primary effect of main magnetic field inhomogeneity. This is of no consequence, however, since the phase information is abandoned at this stage of the processing.

In order to derive an N x N proton density image, N samples from each of N S_A signals must be collected. The N signals have N different distributions of the phase shift along Z and hence N different values of $\int_4 dt\, G_z$. For this purpose a series of waveforms for G_z, namely G_{zo}, G_{z1}, ..., $G_{z(N-1)}$ are utilised such that, for example

$$G_{zo} = 0$$
$$\int_4 dt\ G_{z2} = 2 \int_4 dt\ G_{z1} = 2G^*$$
$$\int_4 dt\ G_{z3} = 3 \int_4 dt\ G_{z1} = 3G^*$$
$$\int_4 dt\ G_{z(N-1)} = (N-1) \int_4 dt\ G_{z1} = (N-1)G^*$$

where $G^* = \int_4 dt\ G_{z1}$.

In other words, the Z gradient is always applied for the same period of time but changes in strength for different pulse sequences. In fact in each successive sequence G_z has the same shape and length but its amplitude changes by equal steps from zero to a maximum value. There is a maximum condition on the series, namely that if the total length of the sample in the Z direction is L_z, then $\gamma L_z G < 2\Pi$. If this limit is exceeded, aliasing will occur, and some parts of the sample may contribute to more than one region of the image.

The image is finally obtained by applying a two-dimensional Fourier transformation to the N echo signals consisting of N samples each. If the signals are designated as $f_m(\tau n)$, where τn are the sampling times and m and n go from 1 to N, an example of such a transformation is given by

$$P(I,J) = \frac{1}{N} \sum_{m=1}^{N} \exp \left[-jm\gamma G^* L_z \left(\frac{2I - N}{2N}\right)\right] \times \sum_{n=1}^{N} \exp \left[-jn\gamma G_x L_x \left(\frac{2J - N}{2N}\right)\right] fm(\tau n) \qquad (3)$$

where L_x is the length of the sample in the X direction, L_z is the length of the sample in the Z direction, P(I,J) is the image element at co-ordinate (I,J), and I and J each go from 1 to N.

An image containing mostly T_1 information can be obtained by collecting N S_B signals along with N S_A signals, deriving an S_A image array and an S_B image array as outlined above, and calculating the T_1 values (from these arrays) corresponding to each imaging element. An image containing a mixture of T_1 information and proton density information can be obtained from a collection of N S_B signals alone.

The method described above lessens the effect of inhomogeneities in the static magnetic field. The effect of an inhomogeneous static field as in equation (1) can be included thus

$$\Delta\phi = \gamma Z_o \int_4 dt\ G_z + \gamma\Delta B(Z_o)\ T' \qquad (4)$$

$\Delta B(Z_o)$ is the difference between the static field value at Z_o and the nominal static field value, and T' is the duration of interval 4. This difference is due to inhomogeneities in the static field. Equation (4) shows that there is extra phase shift caused by the inhomogeneity, but the extra phase shift is the same for all pulse sequences because T' is the same for all pulse sequences. This extra, constant, phase shift for all signals cannot affect the linearity and scale in the Z direction. This conclusion relaxes the stringent condition imposed on the homogeneity of the static magnetic field by the method of U.S. Patent No. 4,070,611.

CLAIMS

1. A method of deriving image information from an object using nuclear magnetic resonance signals comprising subjecting an object to a continuous static magnetic field and carrying out a set of sequential steps which comprise:

1) selectively exciting nuclear spins in a selected plane in the presence of a first gradient of the magnetic field which has a gradient direction perpendicular to said plane;

2) reversing the direction of the said first gradient and applying a second gradient and a third gradient of the magnetic field, the direction of the second gradient being orthogonal to the gradient direction of the first gradient, and the third gradient having a gradient direction orthogonal to the gradient directions of both the first and second gradients;

3) reversing the direction of the second gradient of the magnetic field and holding the said reversed gradient while reading out the resultant free induction decay signal from the said object;

and then successively repeating the above set of steps with different values of gradient of the third gradient, there being a recovery interval between the repetition of successive sets of steps.

2. The method as claimed in claim 1 in which the period of application of the said third gradient of the magnetic field is equal in each of the sets of steps.

3. The method as claimed in claim 1 or claim 2 in which the said first, second and third gradients of the magnetic field in step 2) are all applied simultaneously.

4. The method as claimed in any one of the preceding claims in which the plane in which nuclear spins are selectively excited lies parallel to the direction of the magnetic field.

5. The method as claimed in any one of the preceding claims in which prior to step 1) there is an initial step of inverting the

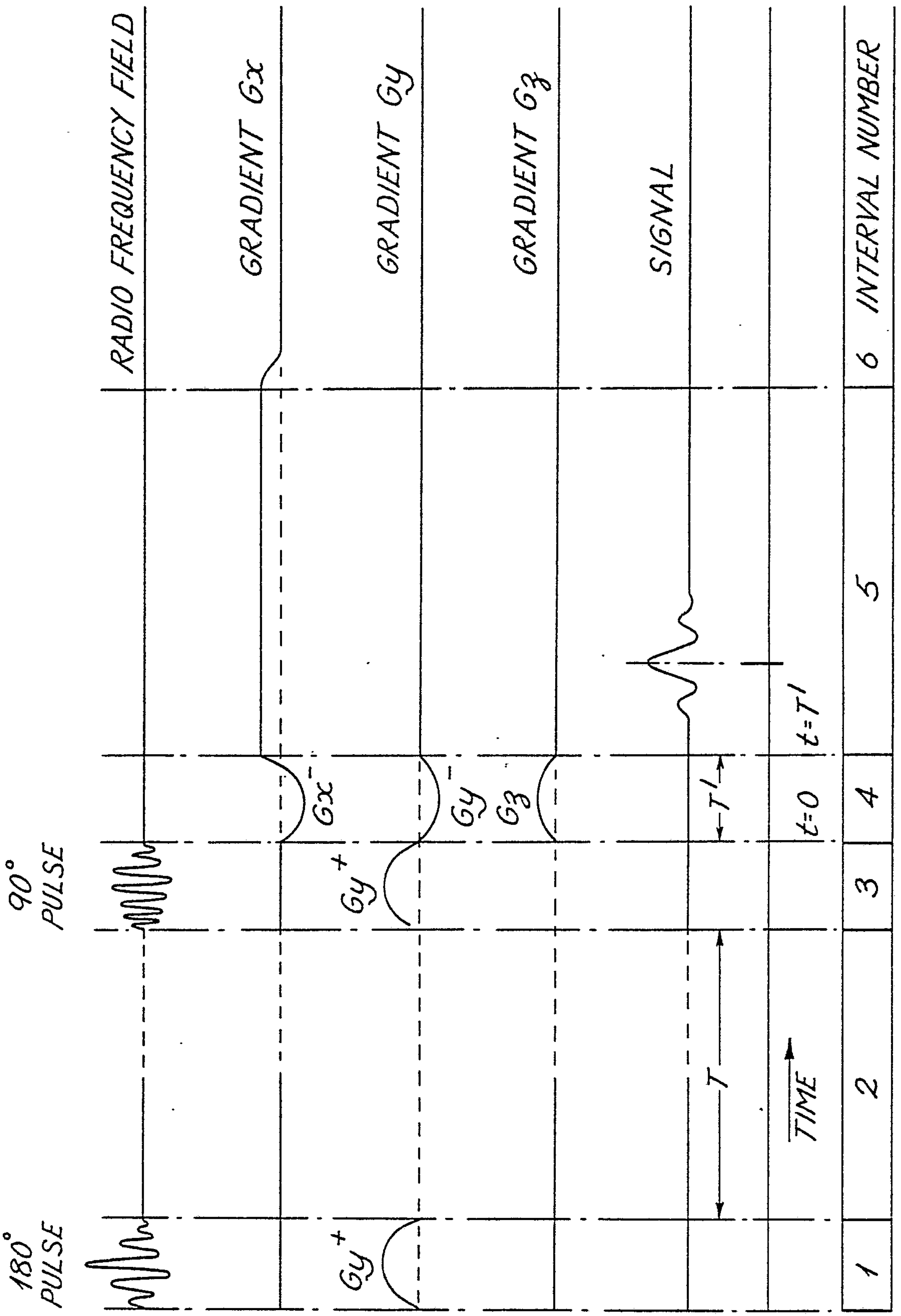
spins followed by an interval approximately equal to the average relaxation time of the spins.

6. The method as claimed in claim 5 in which the said inversion is achieved by an adiabatic fast passage.

7. The method as claimed in any one of the preceding claims in which successive different values of the third gradient differ from each other by equal amounts.

8. The method as claimed in any one of the preceding claims in which the free induction signals are Fourier transformed.

9. A method of deriving image information from an object substantially as described herein with reference to the accompanying drawing.



180° PULSE
90° PULSE
RADIO FREQUENCY FIELD
GRADIENT Gx
GRADIENT Gy
GRADIENT Gz
SIGNAL
INTERVAL NUMBER
Gy+
Gy+
Gx−
Gy−
Gz
T
T'
t=0
t=T'
TIME
1
2
3
4
5
6

INTERNATIONAL SEARCH REPORT

International Application No PCT/GB 81/00044

I. CLASSIFICATION OF SUBJECT MATTER (if several classification symbols apply, indicate all) [3]

According to International Patent Classification (IPC) or to both National Classification and IPC

Int.Cl.[3] G 01 N 24/08

II. FIELDS SEARCHED

Minimum Documentation Searched [4]

Classification System	Classification Symbols
Int.Cl.[3]	G 01 N 24/08; G 01 N 24/00; G 01 N 24/06

Documentation Searched other than Minimum Documentation to the Extent that such Documents are Included in the Fields Searched [5]

III. DOCUMENTS CONSIDERED TO BE RELEVANT [14]

Category *	Citation of Document, [16] with indication, where appropriate, of the relevant passages [17]	Relevant to Claim No. [18]
	DE, A, 2833800, published December 6, 1979 see pages 1,2,8 and figure 1, Hutchison et al.	1,8,9
	DE, A, 2755956, published June 22, 1978 see pages 1-5 and figures 1-6, National Research Development	1,4,8,9
P	Journal of Physics E-Scientific Instruments, vol. 13, no. 9, published September 1980, (London, GB), J.M.S. Hutchison et al.: "A whole-body NMR imaging machine", see pages 947-948 and figure 1	1,8,9

* Special categories of cited documents: [15]

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"X" document of particular relevance

IV. CERTIFICATION

Date of the Actual Completion of the International Search [2]	Date of Mailing of this International Search Report [2]
15th June 1981	7th July 1981

International Searching Authority [1]	Signature of Authorized Officer [20]
EUROPEAN PATENT OFFICE Branch at The Hague P.O.Box 5818 Patentlaan ,2 2280 HV RIJSWIJK (ZH) The Netherlands	G.L.M. Kruydenberg

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