(54) Title of the Invention: Correcting for main magnetic field drift in MRI scanners
Abstract Title: MRI Navigator pulse sequences and drift correction

(57) Two 3D interleaved navigator pulse sequences (echo planar pulse sequences) with low flip angle and different echo times are used between imaging of consecutive voxels. A magnetic field map is obtained by complex division of the resultant pair of navigator images. This is used to obtain parameters to adjust the MRI frequency to compensate for B0 drift and shim to compensate for B0 inhomogeneity. Comparison with reference navigator images can also correct for subject motion. The system central frequency is adjusted by adjusting a phase and frequency for a Numerically Controlled Oscillator (NCO) for all radiofrequency (RF) excitation pulses, and for Analogue-to-Digital converter (ADC) pulses for both the parent scanning sequence and the two navigator pulse sequences.
Figure 1

Figure 2
300  Volume i acquisition

302  Apply first navigator

304  Computer image obtains first navigator image and compares it to a stored reference navigator image to evaluate motion parameters and store motion parameters

306  Apply second navigator

308  Computer image obtains second navigator image and calculates magnetic field map by complex division of first and second navigator images

310  Use field map to determine parameters required to adjust central frequency of the MRI scanner and to adjust homogeneity of the main magnetic field

312  Adjust for subject motion using stored motion parameters

314  Adjust the shim of the MRI scanner

316  Adjust the central system frequency of the MRI scanner

318  Volume i+1 acquisition

Repeat

Figure 3
Figure 4A

Figure 4B
CORRECTING FOR MAIN MAGNETIC FIELD DRIFT IN MRI SCANNERS

FIELD OF THE INVENTION

This invention relates to correcting for main magnetic field (B0) drift in imaging scanners such as Magnetic Resonance Imaging (MRI) scanners.

BACKGROUND TO THE INVENTION

Magnetic Resonance Imaging (MRI) involves applying different types of electromagnetic fields and radiofrequency (RF) excitations to a subject. The aim in doing so is to generate spatial RF signals from a specific region of the subject by which an MRI image is generated. The spatial RF signals are proportional to the strength and the homogeneity of the applied magnetic fields. If there is any distortion in the magnetic fields, this causes distortion in the final MRI image which in some cases may lead to a false diagnosis, for example in medical applications where the subject being imaged is a human subject.

Due to the nature of objects being imaged as well as the difficulties attached to engineering magnetic coils, magnetic fields are not perfectly homogeneous. Factors leading to inhomogeneous effects can include internal factors such as production tolerances in the scanner, heating of coils during scanning, vibrations during scanning, or external factors like ferromagnetic material like iron that may be in the vicinity of the scanner such as in a surrounding building construction.

MRI scanners have a set of several coils, which typically include a main superconductive coil which produces a powerful main magnetic field (called
“B0”) which polarizes an object to be scanned, an RF coil for generating and receiving RF pulses, and magnetic field gradient coils that generate spatial variations in the main magnetic field for scanning a specific region of the subject. In addition, most MRI scanners may include a shim, which is an active or passive device which can adjust the homogeneity of the magnetic field. The shim is used in a process called “shimming” to correct inhomogeneous effects in the magnetic field before scanning starts. In most MRI scanners, the shim is an active shim which includes a number of coils which are provided around the main superconductive coil and which generate small magnetic fields that are superimposed on the main magnetic field to correct inhomogeneous effects.

In existing MRI scanners such as those made by Siemens, Philips or General Electric, shimming of the main magnetic field (B0) is performed once before an MRI pulse sequence begins. While the main magnetic field (B0) is generated by passing a current through the main superconductive coil, small corrective magnetic fields are generated using the additional shim coils.

The fields required to be generated by the shim coils must be determined by first acquiring a map of the main magnetic field. This map can be obtained using various techniques, such as using a two-TE (where TE refers to the echo time) three dimensional gradient echo sequence. From the field map, zero, first and even higher order shims (for example, in Siemens MRI scanners) are calculated and the currents which are required to generate the small magnetic fields in the shim coils are estimated from the field map parameters. These small magnetic fields are then superimposed on the main magnetic field during scanning so as to improve the homogeneity of the main magnetic field.

Some MRI modalities, such as Functional MRI (fMRI) and Diffusion Tensor Imaging (DTI) require scanning a volume of a subject repeatedly using echo planar imaging (EPI) for data acquisition. Such repeated scanning could take
anything between 6 minutes up to about 40 minutes for some DTI applications. During such a long scanning period, the initial shim prepared by the scanner could be compromised, rendering the final MRI images inaccurate. Temporal changes of the initial prepared shimmed main magnetic field may arise due to factors such as patient respiration, poor shimming of the MRI scanner, or generalised and random patient motion. The changes in the main magnetic field, including drift in a system central frequency (zero order shim) and distortion in the shim magnetic field gradients (first or second order shim), can cause different types of geometrical distortions in an MRI image such as shift, stretching, contraction, signal loss, image blurring and ineffective RF excitation pulses.

Currently, the mechanism by which various sources of distortion affect the change in the main magnetic field is not well understood, and existing scanners generally do not include an ability to compensate for such changes during the course of scanning. Frequency drift of the system central frequency, in particular, cannot be addressed with external tracking systems.

The technology described in this application seeks to address these problems, at least to some extent.

**SUMMARY OF THE INVENTION**

According to the invention there is provided a method of correcting for main magnetic field (B0) drift in a Magnetic Resonance Imaging (MRI) scanner during a scanning sequence which includes the acquisition of successive volumes by means of Magnetic Resonance (MR) pulse sequences, comprising:

- interleaving a first three-dimensional navigator pulse sequence into the scanning sequence by applying a first three-dimensional navigator after the acquisition of each volume in the scanning sequence;
- interleaving a second three-dimensional navigator pulse sequence into
the scanning sequence by applying a second three-dimensional navigator
after each first navigator and before the acquisition of the next volume in the
scanning sequence, wherein the first and second navigator sequences have
different echo times; and

after each pair of first and second navigators have been applied:
  obtaining resulting first and second navigator images;
  determining a magnetic field map by complex division of the
  first and second navigator images;
  using the magnetic field map to determine parameters required
to adjust a system central frequency of the MRI scanner to
compensate for a drift in the main magnetic field (B0); and
  adjusting the system central frequency of the MRI scanner
  based on the determined parameters.

Further features provide for the determined parameters to include a zero
order shim which is estimated from the magnetic field map based on a least-
squares fit and used to determine the parameters for adjusting the central
frequency.

Still further features provide for the system central frequency of the MRI
scanner to be adjusted by adjusting a phase and frequency for a Numerically
Controlled Oscillator (NCO) for the MRI scanner for all radiofrequency (RF)
excitation pulses of the MRI scanner and for Analogue to Digital Converter
(ADC) pulses for both the scanning sequence and the two navigator pulse
sequences.

Yet further features provide for the method to include:
  using the field map to determine parameters required to adjust the
  homogeneity of the main magnetic field of the MRI scanner, and
  adjusting a shim coil of the MRI scanner with the determined
  parameters before acquisition of the next volume in the scanning sequence.
Further features provide for the determined parameters to include first order linear shims which are estimated from the magnetic field map and used to adjust the distortion in the magnetic field gradients of the shim coil.

Still further features provide for the method to include, after obtaining each first navigator image, estimating motion parameters which include three translations and three rotations, and updating for motion after the acquisition of the second navigator and before acquisition of the next volume in the scanning sequence.

Yet further features provide for the first and second three dimensional navigator sequences to be echo planar imaging (EPI) navigator sequences; for the EPI navigator sequences to be identical sequences except for having different echo times; and for echo times to be chosen for the first EPI navigator and second EPI navigator so that a resultant signal produced by the excitation of fat and water by the two EPI navigators is in phase, so that any phase evolution between the two navigators is not affected by the phase difference of fat and water. In one embodiment, an echo time of 6.6 milliseconds is chosen for the first EPI navigator and an echo time of 9 milliseconds is chosen for the second EPI navigator so that the resultant signal produced by the excitation of fat and water is in phase at main magnetic (B0) field strength of 3 Tesla.

Further features provide for the scanning sequence to be one of: a Functional Magnetic Resonance Imaging (fMRI) scanning sequence, and a Diffusion Tensor Imaging (DTI) scanning sequence.

Still further features provide for the navigators to have a low flip angle to ensure that an MRI contrast of each navigator does not influence a contrast of the MRI scanning sequence. In one embodiment, a flip angle of about 2° is chosen.
Yet further features provide for the first and second EPI navigator pulse sequences to have a sufficiently low spatial resolution so that the navigator images can be obtained in sufficient time. For example, the navigator sequences may have a spatial resolution of 8 x 8 x 8 mm$^3$ in which case the navigator images can be obtained in approximately 0.475 seconds.

**BRIEF DESCRIPTION OF THE DRAWINGS**

In the drawings:-

- **Figure 1** is a block diagram of hardware and logical components of a Magnetic Resonance Imaging (MRI) scanner;

- **Figure 2** is a timing diagram of a scanning sequence in which magnetic field inhomogeneity is corrected;

- **Figure 3** is a flow diagram showing how corrections for magnetic field drift, magnetic field inhomogeneity and subject motion can be achieved with the scanning sequence shown in Figure 2;

- **Figure 4A** is a graph showing manual changes in scanner central frequency and measured changes in one experiment; and

- **Figure 4B** is a graph showing measured linear shim gradients following manual distortion of a static shim.

**DETAILED DESCRIPTION WITH REFERENCE TO THE DRAWINGS**

Figure 1 is a block diagram of various hardware and logical components of a Magnetic Resonance Imaging (MRI) scanner (100) used in the invention. The MRI scanner includes a main superconductive coil (102) which produces a powerful main magnetic field, known as “B0”. A radiofrequency (RF) coil
(104) is provided for generating and receiving RF pulses, and magnetic field
gradient coils (106) generate spatial variations in the main magnetic field (B0)
for scanning a specific region of interest. Active shim coils (108) are also
provided, which are additional coils provided around the main
superconductive coil (102) which generate small magnetic fields that are
superimposed on the main magnetic field (B0) to correct inhomogeneous
effects.

The MRI scanner (100) also includes a Numerically Controlled Oscillator
(NCO) (110) which is used to adjust a frequency and phase for all RF
excitation pulses and the resultant received RF signals. Because of distortion
or drift in the main magnetic field (B0) or as a result of subject motion, the
phase and frequency of the excitation pulses generated by the RF coil (104)
may change, so the NCO is used to adjust the phase and frequency of the
RF excitation pulses.

The resultant RF signal received from the subject must be digitized before it
can be analysed and an image constructed therefrom, and an Analogue-to-
Digital Converter (ADC) (112) is provided for digitizing received analogue RF
signals. The phase and frequency of the resultant RF signal may change due
to inhomogeneity, and therefore the phase and frequency of NCO (110) of
the ADC (112) has to be corrected. The ADC feeds into a computer image
(114), which is a logical component that includes all processes and programs
related to data acquisition and image calculation and reconstruction.

The MRI scanner (100) is configured by using an MRI sequence component
(116), which is a software component that by means of which the MRI
scanner (100) can be controlled. For example, a radiographer can prepare a
specific MRI protocol by using the MRI sequence component (116). The
prepared MRI protocol sets out the specific type, duration and other
parameters of an MRI scanning sequence, which is then applied to the other
hardware components such as the RF coil (104), magnetic field gradient coils
(106), and NCO (110) to conduct a scanning sequence.

Figure 2 shows a timing diagram of a scanning sequence (200) in which main magnetic field (B0) drift is corrected according to the technology. The illustrated pulse sequence is written in an object oriented program (OOP) and loaded onto the MRI scanner (100). In this illustration, the scanning sequence (200) includes a parent sequence (202) which is either a functional MRI (fMRI) or Diffusion Tensor Imaging (DTI) scanning sequence that includes acquisition of successive volumes of a subject being imaged. In this illustrated portion of the timing diagram, the acquisition of three volumes is shown, a first volume (202a), second volume (202b) and third volume (202c), which are successively acquired by the MRI scanner. These volumes could, for example, be imaged volumes in a human subject such as small portions of the human brain.

The scanning sequence also includes a first navigator pulse sequence (204) which has been interleaved into the parent sequence (202). The first navigator pulse sequence (204) includes a series of navigators which are applied directly after each volume acquisition (202a, 202b, 202c). In this illustration, two of the navigators (204a, 204b) in the first navigator pulse sequence (204) are shown.

The scanning sequence (200) also includes a second navigator pulse sequence (206) which has been interleaved into the parent sequence (202). The second navigator pulse sequence (206) includes a series of navigators which are applied after each first navigator and before the acquisition of the next volume in the parent scanning sequence (202). In this illustration, two of the navigators (206a) in the second navigator pulse sequence (206) are shown.

The first and second navigator pulse sequences (204, 206) are three dimensional navigator echo planar imaging (EPI) pulse sequences. Being
three dimensional, the navigators are volumetric as opposed to linear navigators. The navigator pulse sequences are identical except that they have different echo times. An echo time is the time between the applied RF excitation pulse and the peak of the resultant echo signal. In one experiment, an echo time of 6.6 milliseconds was chosen for the first navigator and an echo time of 9 milliseconds was chosen for the second navigator. For these echo times, any phase evolution between the two navigators is not affected by the phase difference of fat and water in the imaged subject, but the phase evolution is instead accumulated due to main magnetic field (B0) inhomogeneity.

The navigators are chosen to have a low flip angle to ensure that an MRI contrast of the navigators do not influence a contrast of the parent sequence (202). In one experiment, a flip angle of 2° was chosen. The navigator pulse sequences are also chosen to have a sufficiently low spatial resolution so that resultant navigator images can be obtained in a sufficiently short space of time. In one experiment, the navigator sequences may have spatial resolutions of 8 x 8 x 8 mm³ in which case the navigator images can be obtained in approximately 0.475 seconds.

For each navigator, the computer image (114) obtains a resultant navigator image. During scanning, the first navigator image is affected by a field inhomogeneity of a phase of $\Delta \Phi t1$, and the second navigator image is affected by different phase $\Delta \Phi t1 + \tau$. Because the two navigators share a phase of $\Delta \Phi t1$, a three-dimensional field map $\Delta \Phi t$ can be obtained by complex division of the two navigators' images. Complex division is the division of two complex numbers, which can be done according to known techniques by multiplying the numerator and the denominator by the complex conjugate of the denominator.

The three dimensional field map $\Delta \Phi t$ is directly proportional to the main magnetic field inhomogeneity. From the main field map, shimming
parameters with different orders can be estimated. The determined parameters include a first or even second order shim which is estimated from the magnetic field map based on a least-squares fit. The shimming parameters are those parameters required to adjust the homogeneity of the main magnetic field including first order linear shims. The required currents for optimizing the shim gradient coils are then obtained from the shim parameters according to known techniques and the active shimming coils activated so as to adjust the homogeneity of the main magnetic field. Typically, calculated shim offsets are added to initial shims that are prepared by the MRI scanner.

The three dimensional field map $\Delta \Phi_\tau$ is also used to determine parameters required to adjust a system central frequency of the MRI scanner to compensate for a drift in the main magnetic field (B0). The determined parameters include a zero order shim which is estimated from the magnetic field map based on a least-squares fit and used to determine the parameters for adjusting the central frequency. Adjustment of the central frequency can be done by adjusting the phase and frequency for the Numerically Controlled Oscillator (110) of the MRI scanner for all RF excitation pulses of the MRI scanner and for the analogue to digital converter (112) pulses for both the scanning sequence and the two navigator sequences. Adjusting the analogue to digital converter is done by adding an offset phase to each line of K-space.

Motion correction is also affected by using the first navigator pulse sequence (204) and comparing each navigator image with a previously stored navigator image, as will be further described herein.

Figure 3 is a flow diagram that shows successive stages carried out in a method of correcting for main magnetic field (B0) drift and also correcting for the distortion in the magnetic field shim gradients and subject motion, and is to be understood with reference to the timing diagram of Figure 2 and the
block diagram of Figure 1.

At a first stage (300), the first volume i (202a) of DTI or fMRI is acquired during a pulse sequence that is part of the parent sequence (202). At a next stage (302), the first navigator (204a) is applied immediately after the first volume acquisition (202a). At a next stage (304), the computer image (114) obtains a first navigator image and compares it to a stored reference navigator image. By comparing these images, the computer image (114) is able to obtain motion parameters that include three translations and three rotations, which are the parameters required to calculate both translation and motion in a three axis (x, y and z) coordinate system. Motion calculation is done by the computer image (114) and these motion parameters are then sent and stored in the MRI sequence component (116). In one experiment, the required time for motion calculation by the computer image (114) and delivery to the sequence component (116) was approximately 100 milliseconds.

At a next stage (306), the second navigator (206a) is then applied. At a next stage (308), the computer image (114) obtains the second navigator image and calculates a magnetic field map by complex division of the first and second navigator images as previously explained.

The field map is then used, at a next stage (310), to determine parameters that are required to adjust the central system frequency of the MRI scanner and the homogeneity of the main magnetic field (B0). The computer image (114) then sends and stores these parameters in the MRI sequence component (116). The required time for calculation of the parameters and delivery to the MRI sequence component in one experiment was approximately 80 milliseconds.

At a next stage (312), and before shim correction starts, the MRI sequence component (116) adjusts for subject motion using the stored motion
parameters. This is done by reorienting the scanner coordinate system and accordingly all gradients are reoriented to the correct position and also all RF and ADC pulses through the NCO are corrected for the measured shift in the position of the subject.

Following motion correction, at a next stage (314), the shim of the MRI scanner (100) is then adjusted as previously described to compensate for the inhomogeneity in the main magnetic field (B0), and at a next stage (316) the central frequency of MRI scanner is adjusted by adjusting a phase and frequency of the NCO (110) for all RF excitation pulses and the analogue to digital converter (ADC) pulses of the MRI scanner for both the parent sequence (202) and the two navigator sequences (204, 206).

In this way, the next volume i+1 acquisition (202b), which is shown at a next stage (318), is done with corrections for all three of: central frequency drift, magnetic field inhomogeneity and subject motion. The process then repeats itself. The required time for the error correction in one experiment was 1,150 milliseconds, and this time could be shortened further by using a lower resolution for the navigators or faster processing of the algorithms.

**Experimental results**

The disclosed technique was applied on an Allegra 3T MRI scanner manufactured by Siemens. A stationary water phantom was first scanned to validate the accuracy of the technique in evaluating the drift in the B0 field and the distortion in the shim magnetic field gradients. The system frequency initially prepared by the MRI scanner before acquisition was manually offset in 6 different scans by 5, 10, 20, 40, 70 and 100 Hz respectively. A static shim initially prepared by the scanner before the acquisition sequence was manually adjusted by 15 μT/m in the x direction, 15 μT/m in both the x and y direction, and finally 15 μT/m in x, y and z directions. A zero order shim was calculated to obtain the drift in the B0 field, and first order linear shim
gradients (Gx, Gy, and Gz) were obtained to determine inhomogeneous effects.

Figure 4A is a graph that shows the ability of the disclosed technique (DvNav) to evaluate the manual changes that were made to the scanner central frequency. As can be seen, when the frequency is manually changed, the DvNav sequence accurately measures those changes. The technique is thus effective in measuring and correcting changes in the scanner central frequency that may result from, for example, heating of the iron comprising the shim coils.

Figure 4B is a graph that shows the ability of the disclosed technique (DvNav) to evaluate changes made in the linear shim gradients in x, then x and y, and finally x, y and z directions. As can be seen, where the linear shim gradients are manually changed, the DvNav sequence accurately measures those changes. The technique is thus effective in measuring and correcting changes required to adjust linear shims to compensate for in homogeneities in the main magnetic field (B0).

In experiments on human subjects, it was found that subject motion alters the initial shim and can produce a significant frequency drift which can cause a shift of approximately 3mm (which equates to 30 Hz) in the image space. The techniques described herein are able to simultaneously correct for both subject motion and drift in the main magnetic field (B0), which cannot be addressed with previously disclosed external tracking systems or prospective or retrospective motion correction. Another advantage of the disclosed technique is the ability to shim over a specific region slab-by-slab or slice-by slice fashion, which is very important when scanning a small region.

The described technique is able to measure, report and correct in real time for changes in the main magnetic field, changes in first order shims and subject motion simultaneously. The correlation between subject motion,
motion correction, the drift in the scanner central frequency and first order shims can also be studied. Higher order shims can be implemented if the hardware of the MRI scanner allows for this.
CLAIMS:

1. A method of correcting for main magnetic field (B0) drift in a Magnetic Resonance Imaging (MRI) scanner during a scanning sequence which includes the acquisition of successive volumes by means of Magnetic Resonance (MR) pulse sequences, comprising:
   interleaving a first three-dimensional navigator pulse sequence into the scanning sequence by applying a first three-dimensional navigator after the acquisition of each volume in the scanning sequence;
   interleaving a second three-dimensional navigator pulse sequence into the scanning sequence by applying a second three-dimensional navigator after each first navigator and before the acquisition of the next volume in the scanning sequence, wherein the first and second navigator sequences have different echo times; and
   after each pair of first and second navigators have been applied:
      obtaining resulting first and second navigator images;
      determining a magnetic field map by complex division of the first and second navigator images;
      using the magnetic field map to determine parameters required to adjust a system central frequency of the MRI scanner to compensate for a drift in the main magnetic field (B0); and
      adjusting the system central frequency of the MRI scanner based on the determined parameters.

2. The method as claimed in claim 1, wherein the determined parameters include a zero order shim which is estimated from the magnetic field map based on a least-squares fit.

3. The method as claimed in claim 1 or claim 2, wherein the system
The central frequency of the MRI scanner is adjusted by adjusting a phase and frequency for a Numerically Controlled Oscillator (NCO) of the MRI scanner for all radiofrequency (RF) excitation pulses of the MRI scanner and for Analogue to Digital Converter (ADC) pulses of the MRI scanner for both the scanning sequence and the two navigator pulse sequences.

4. The method as claimed in any one of the preceding claims, which includes:
   using the field map to determine parameters required to adjust the homogeneity of the main magnetic field of the MRI scanner, and
   adjusting a shim coil of the MRI scanner with the determined parameters before acquisition of the next volume in the scanning sequence.

5. The method as claimed in claim 4, wherein the determined parameters include first order linear shims which are estimated from the magnetic field map and used to adjust distortion in the magnetic field gradients of the shim coil of the MRI scanner.

6. The method as claimed in any one of the preceding claims, which includes:
   after obtaining each first navigator image, estimating motion parameters which include three translations and three rotations; and
   updating for motion after the acquisition of the second navigator and before acquisition of the next volume in the scanning sequence.

7. The method as claimed in any one of the preceding claims, wherein the first and second three dimensional navigator sequences are echo planar imaging (EPI) navigator sequences, and where the EPI navigator sequences are identical sequences except for having different echo times.
8. The method as claimed in claim 7, wherein the echo times are chosen for the first EPI navigator and second EPI navigator so that a phase difference of the field map between the first and second navigators is not affected by a phase offset of fat or water but rather by an inhomogeneity in the main magnetic field.

9. The method as claimed in claim 8, wherein echo times of about 6.6 milliseconds and about 9 milliseconds are chosen so that a resultant signal produced by the excitation of fat and water by the two EPI navigators is in phase at a main magnetic field strength of about 3 Tesla.

10. The method as claimed in any one of the preceding claims, wherein the scanning sequence is one of: a Functional Magnetic Resonance Imaging (fMRI) scanning sequence, and a Diffusion Tensor Imaging (DTI) scanning sequence.

11. The method as claimed in any one of the preceding claims, wherein the navigators have sufficiently low flip angles to ensure that an MRI contrast of the navigators do not influence a contrast of the MRI scanning sequence.

12. The method as claimed in claim 11, wherein the flip angle is equal to about 2°.

13. The method as claimed in any one of the preceding claims, wherein the first and second navigator pulse sequences have a sufficiently low spatial resolution so that the navigator images can be obtained in a sufficient amount of time.

14. The method as claimed in claim 13, wherein the first and second

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navigator pulse sequences have a spatial resolution of approximately 8 x 8 x 8 mm$^3$ and the navigator images are obtained in approximately 0.475 seconds.
Patents Act 1977: Search Report under Section 17

Documents considered to be relevant:

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<tr>
<td>X</td>
<td>1 - 14</td>
<td>NMR in Biomedicine (2012) vol. 25, Hess et al., &quot;Real-time motion and B0 correction for localized adiabatic selective refocusing (LASER) MRSI using echo planar imaging volumetric navigators&quot;, pp347-358 noting abstract, &quot;LASER sequence&quot; section from p348, figures 1 and 3 especially.</td>
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<td>Magnetic Resonance in Medicine (2011) vol. 66, Hess et al., &quot;Real-time motion and B0 corrected single voxel spectroscopy using volumetric navigators&quot;, pp314-323 see abstract, column 1 of page 314, &quot;The EPI vNAV&quot; section from p316 and &quot;discussion&quot; section on p322 especially</td>
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<td>X</td>
<td>1 - 8, 10 - 14</td>
<td>NeuroImage (2014), vol. 88, Bogner et al., &quot;Real-time motion- and B0-correction for LASER-localized spiral-accelerated 3D-MRSI of the brain at 3T&quot;, pp22 - 31 see especially abstract, &quot;volumetric navigator&quot; section from p23 and &quot;scanner frequency drift correction&quot; sections p25 and p26 especially</td>
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<td>-</td>
<td>US6472872 (MAYO FOUNDATION) see abstract, columns 5 (lines 30 - 58), 6 (line 31) - 7 (line 6), and 7 (line 46) - 9 (line 67), 10 (lines 11 - 13) especially</td>
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G01N; G01R

The following online and other databases have been used in the preparation of this search report

EPODOC, WPI, TXTE, INSPEC, XPIOOP, XPAIP, XPIEE, XPI3E, XPESP, XPESP2, XPSPRNG
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