US007508212B2

(12) **United States Patent**
Fain et al.

(10) **Patent No.: US 7,508,212 B2**
(45) **Date of Patent: Mar. 24, 2009**

(54) **RF COIL ASSEMBLY AND METHOD FOR PRACTICING MAGNETIZATION TRANSFER ON MAGNETIC RESONANCE IMAGING AND SPECTROSCOPY SYSTEMS**

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(*) Notice: Subject to any disclaimer, the term of this patent is extended or adjusted under 35 U.S.C. 154(b) by 0 days.

(21) Appl. No.: **11/841,201**

(22) Filed: **Aug. 20, 2007**

(65) **Prior Publication Data**

US 2008/0231281 A1 Sep. 25, 2008

Related U.S. Application Data

(63) Continuation-in-part of application No. 11/726,643, filed on Mar. 22, 2007.

(51) **Int. Cl.**
G01V 3/00 (2006.01)

(52) **U.S. Cl.** **324/318**; 324/322

(58) **Field of Classification Search** 324/318, 324/322

See application file for complete search history.

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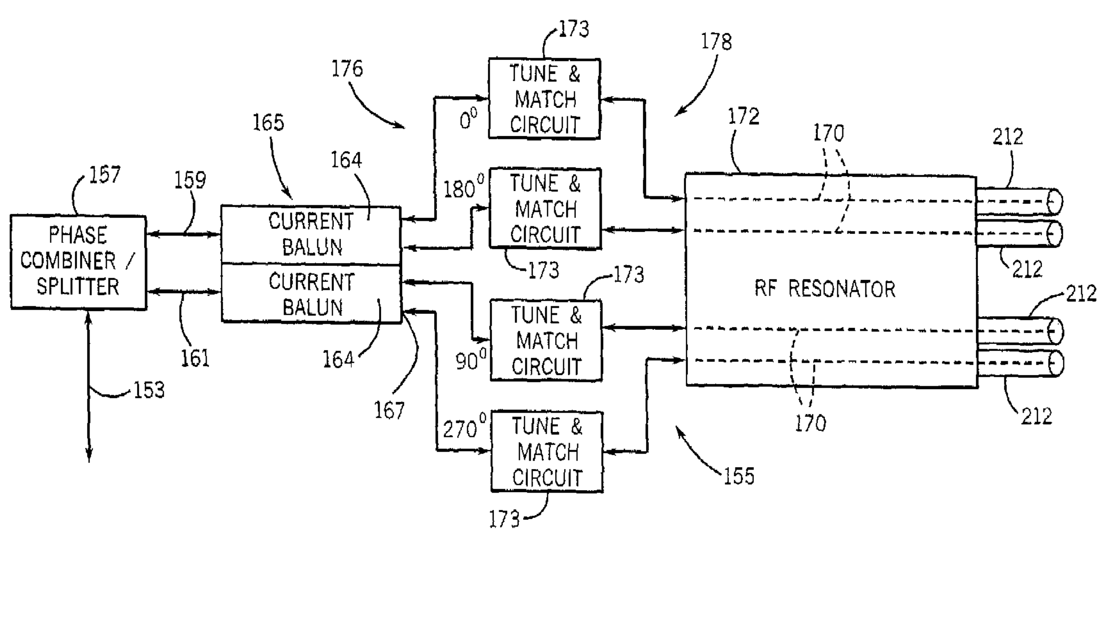
(57) **ABSTRACT**

An RF coil assembly for an MRI system includes a resonator formed by a cylindrical shield and pairs of opposing conductive legs disposed symmetrically around a central axis and extending the axial length of the shield. One set of conductive leg pairs is tuned to operate at the Larmor frequency of ^{13}C and another set is tuned to operate at the Larmor frequency of ^{1}H. Drive circuitry operates the RF coil assembly to produce ^{1}H spin magnetization which is transferred to ^{13}C magnetization by the nuclear overhauser effect and to acquire MR data from the ^{13}C spins. Multinuclear measurements can be made simultaneously at different Larmor frequencies.

10 Claims, 11 Drawing Sheets



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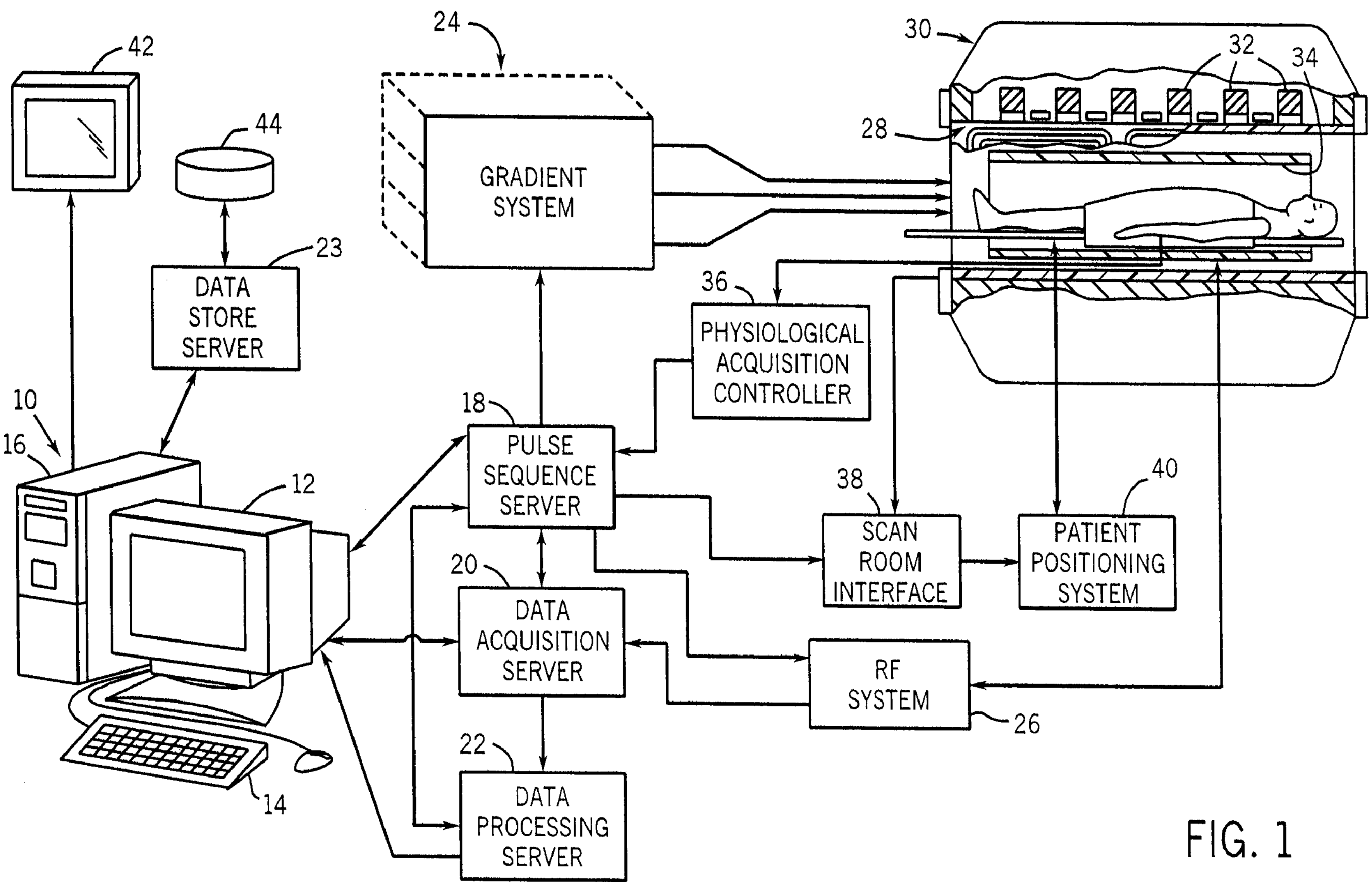

42
44
23
DATA STORE SERVER
24
GRADIENT SYSTEM
30
32
34
28
36
PHYSIOLOGICAL ACQUISITION CONTROLLER
10
16
12
14
18
PULSE SEQUENCE SERVER
20
DATA ACQUISITION SERVER
22
DATA PROCESSING SERVER
38
SCAN ROOM INTERFACE
40
PATIENT POSITIONING SYSTEM
RF SYSTEM
26

FIG. 1

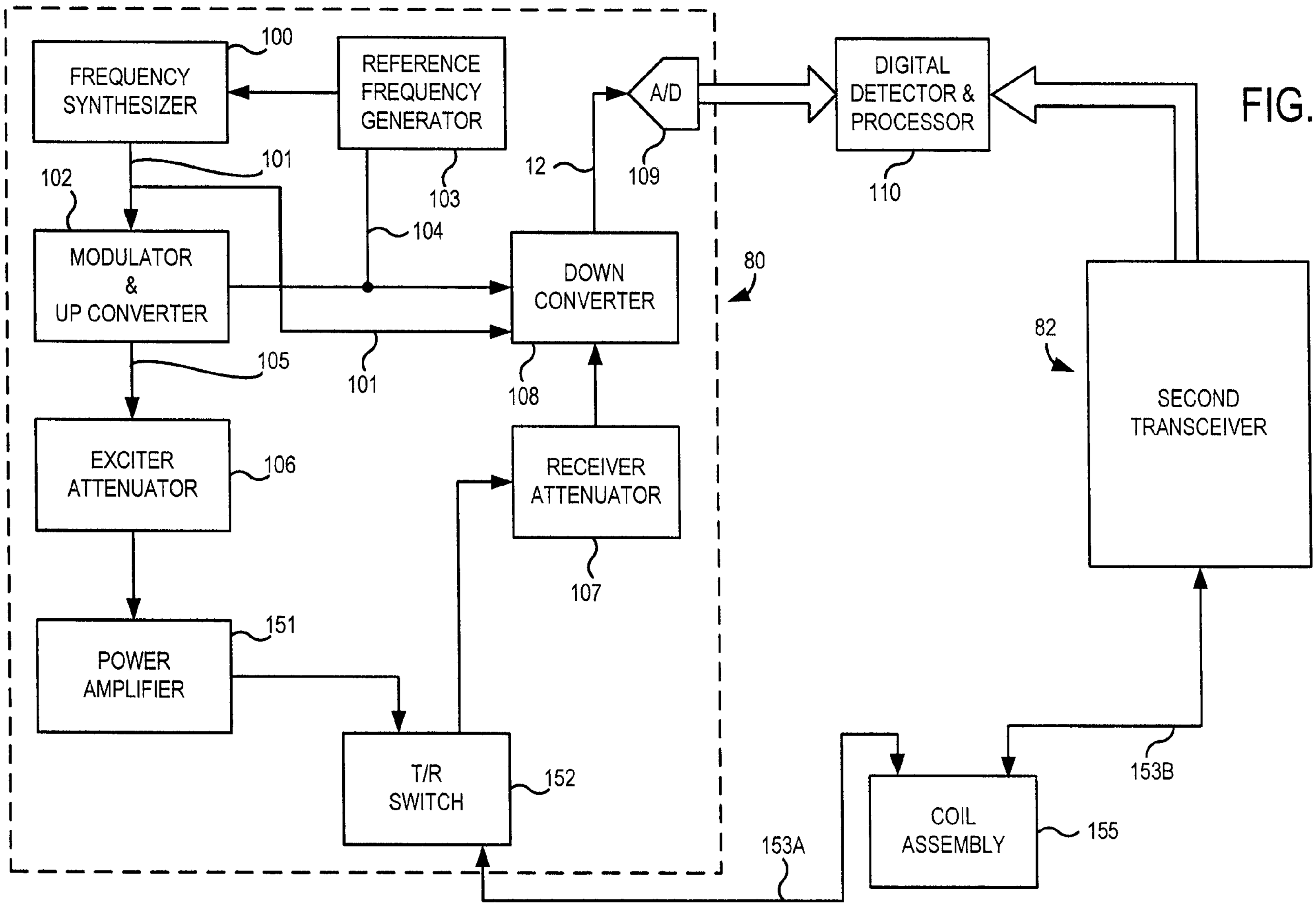

FREQUENCY SYNTHESIZER
100
REFERENCE FREQUENCY GENERATOR
103
A/D
109
12
DIGITAL DETECTOR & PROCESSOR
110
101
102
104
MODULATOR & UP CONVERTER
DOWN CONVERTER
80
105
101
108
82
EXCITER ATTENUATOR
106
RECEIVER ATTENUATOR
107
SECOND TRANSCEIVER
POWER AMPLIFIER
151
T/R SWITCH
152
153A
153B
COIL ASSEMBLY
155

FIG. 2

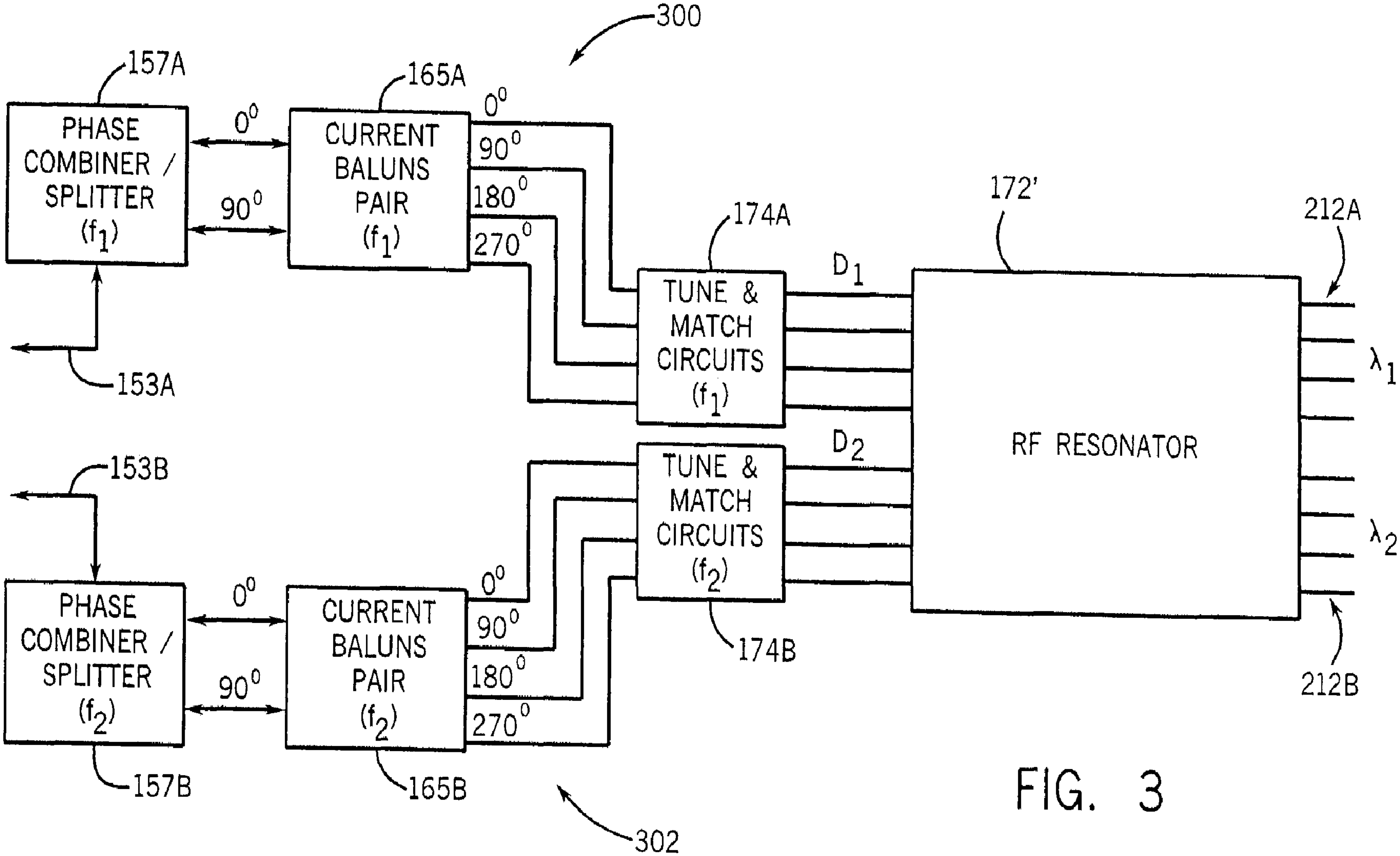

300
157A
165A
PHASE COMBINER / SPLITTER (f_1)
0^0
90^0
CURRENT BALUNS PAIR (f_1)
0^0
90^0
180^0
270^0
174A
TUNE & MATCH CIRCUITS (f_1)
D_1
172'
RF RESONATOR
212A
λ_1
153A
153B
D_2
TUNE & MATCH CIRCUITS (f_2)
174B
PHASE COMBINER / SPLITTER (f_2)
0^0
90^0
CURRENT BALUNS PAIR (f_2)
0^0
90^0
180^0
270^0
λ_2
212B
157B
165B
302

FIG. 3

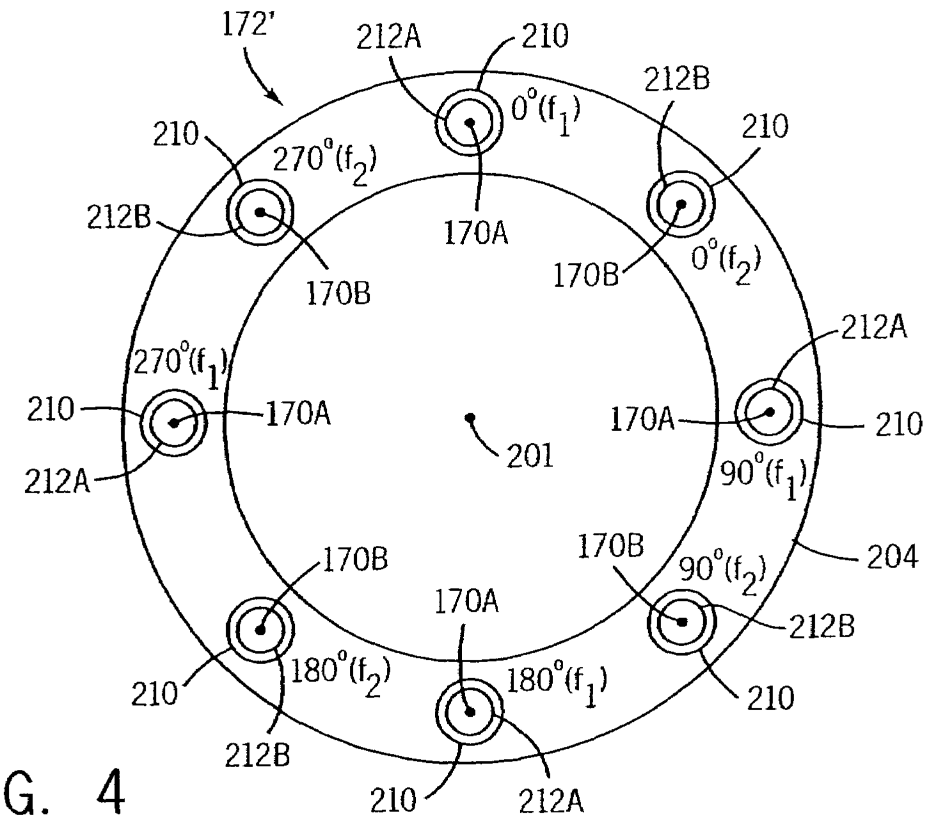

172'
212A
210
0°(f1)
212B
210
210
270°(f2)
212B
170A
170B
0°(f2)
170B
212A
270°(f1)
210
170A
170A
201
210
212A
90°(f1)
204
170B
170B
90°(f2)
170A
212B
210
180°(f2)
180°(f1)
210
212B
210
212A

FIG. 4

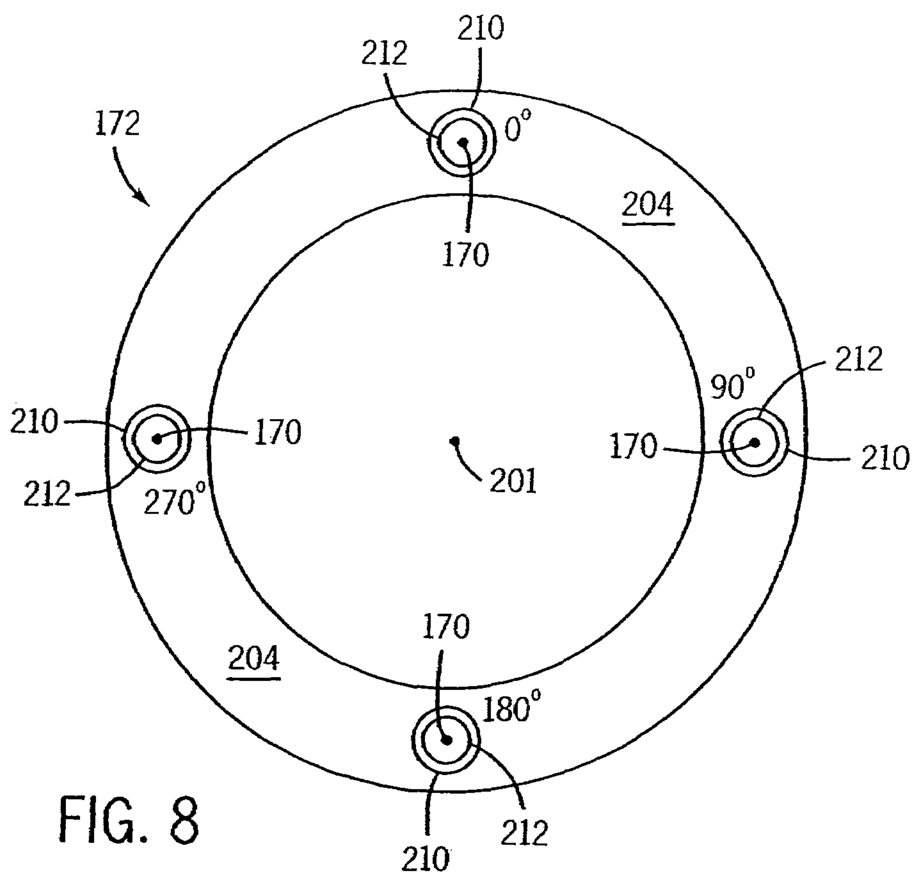

210
212
0°
172
204
170
212
90°
210
170
170
210
201
212
270°
170
204
180°
210
212

FIG. 8

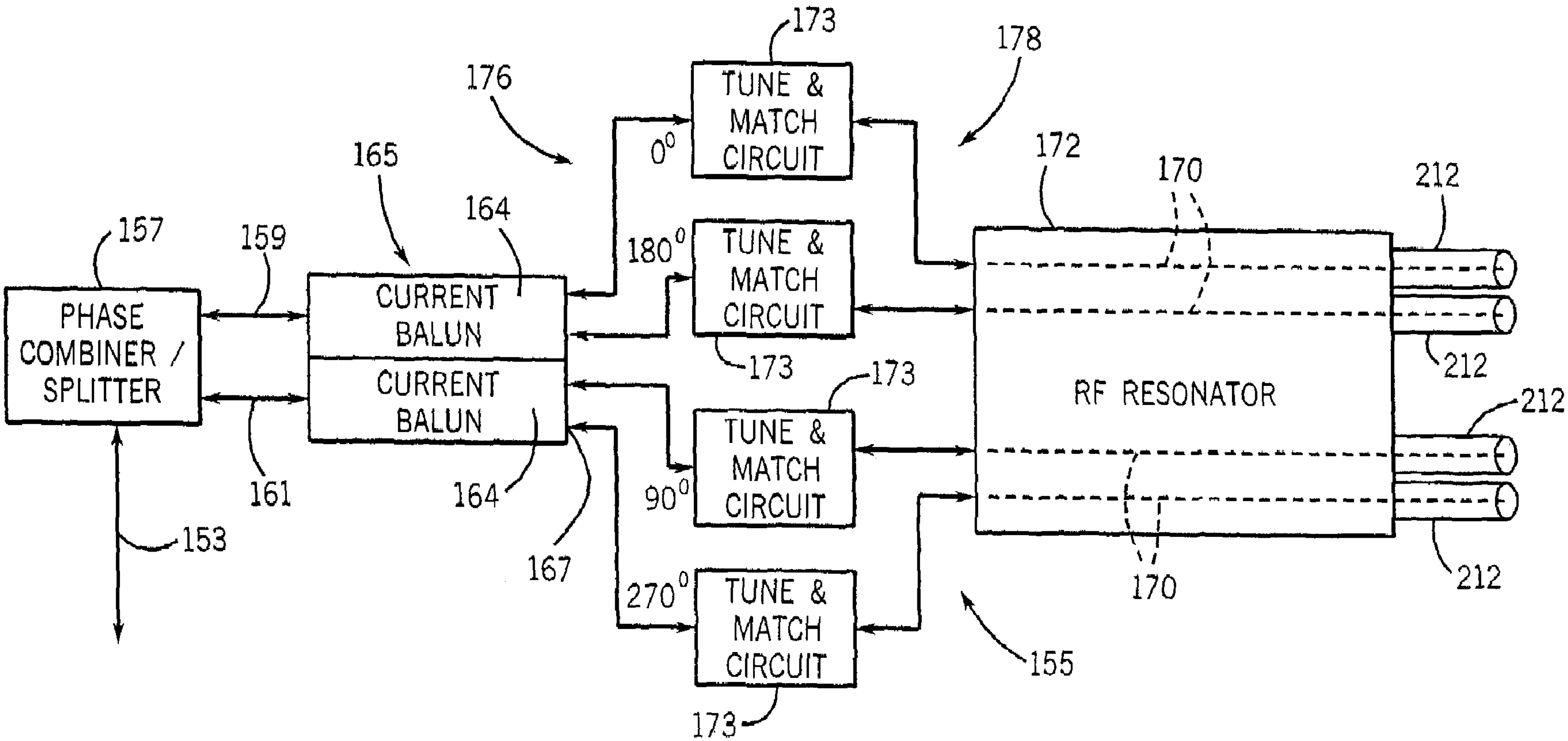

PHASE COMBINER / SPLITTER
CURRENT BALUN
CURRENT BALUN
TUNE & MATCH CIRCUIT
TUNE & MATCH CIRCUIT
TUNE & MATCH CIRCUIT
TUNE & MATCH CIRCUIT
RF RESONATOR
0°
180°
90°
270°
153
155
157
159
161
164
164
165
167
170
170
172
173
173
173
173
176
178
212
212
212
212

FIG. 5

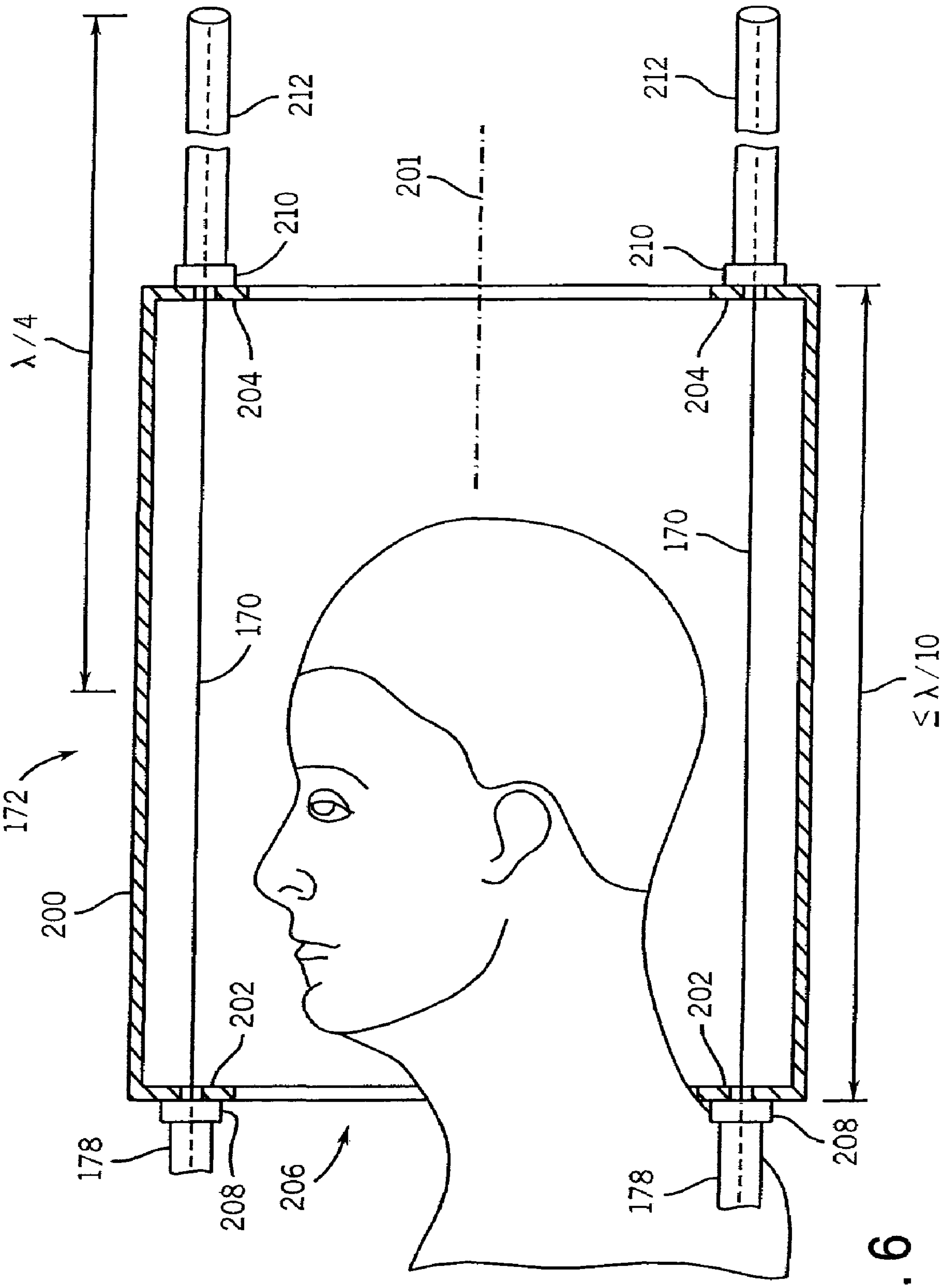
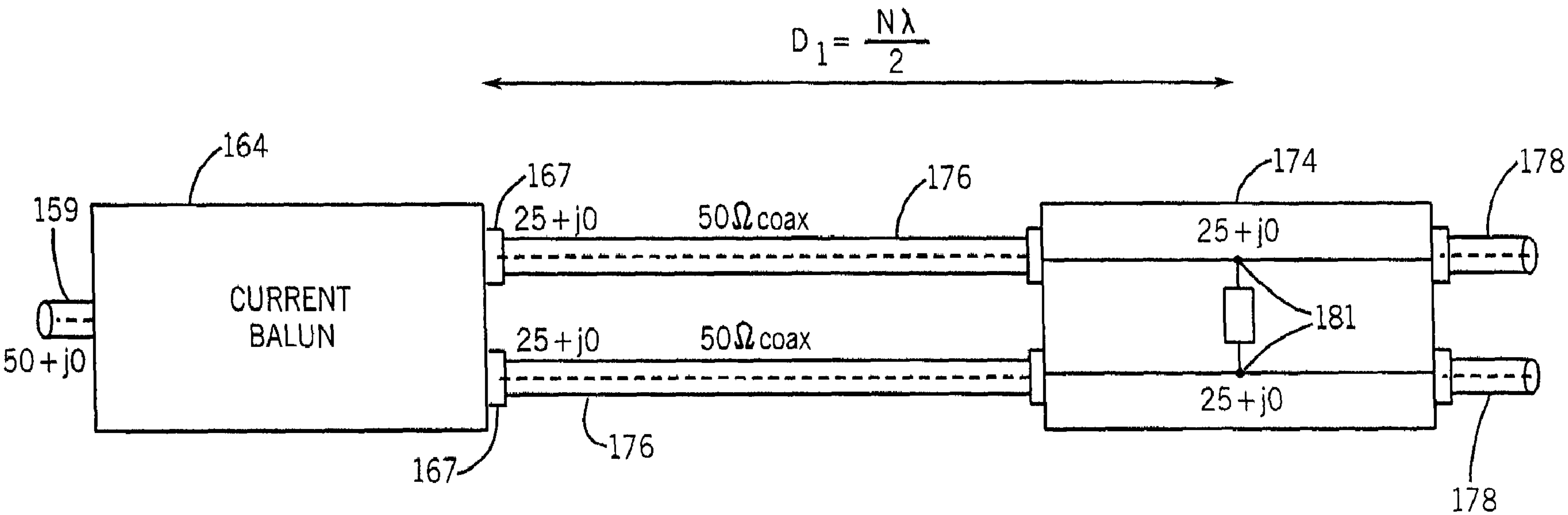

212
212
201
210
210
204
204
λ/4
170
170
172
200
202
202
208
208
178
178
206
≤ λ/10

FIG. 6



$D_1 = \frac{N\lambda}{2}$
164
159
CURRENT BALUN
50 + j0
167
25 + j0
50Ω coax
176
174
178
181

FIG. 7

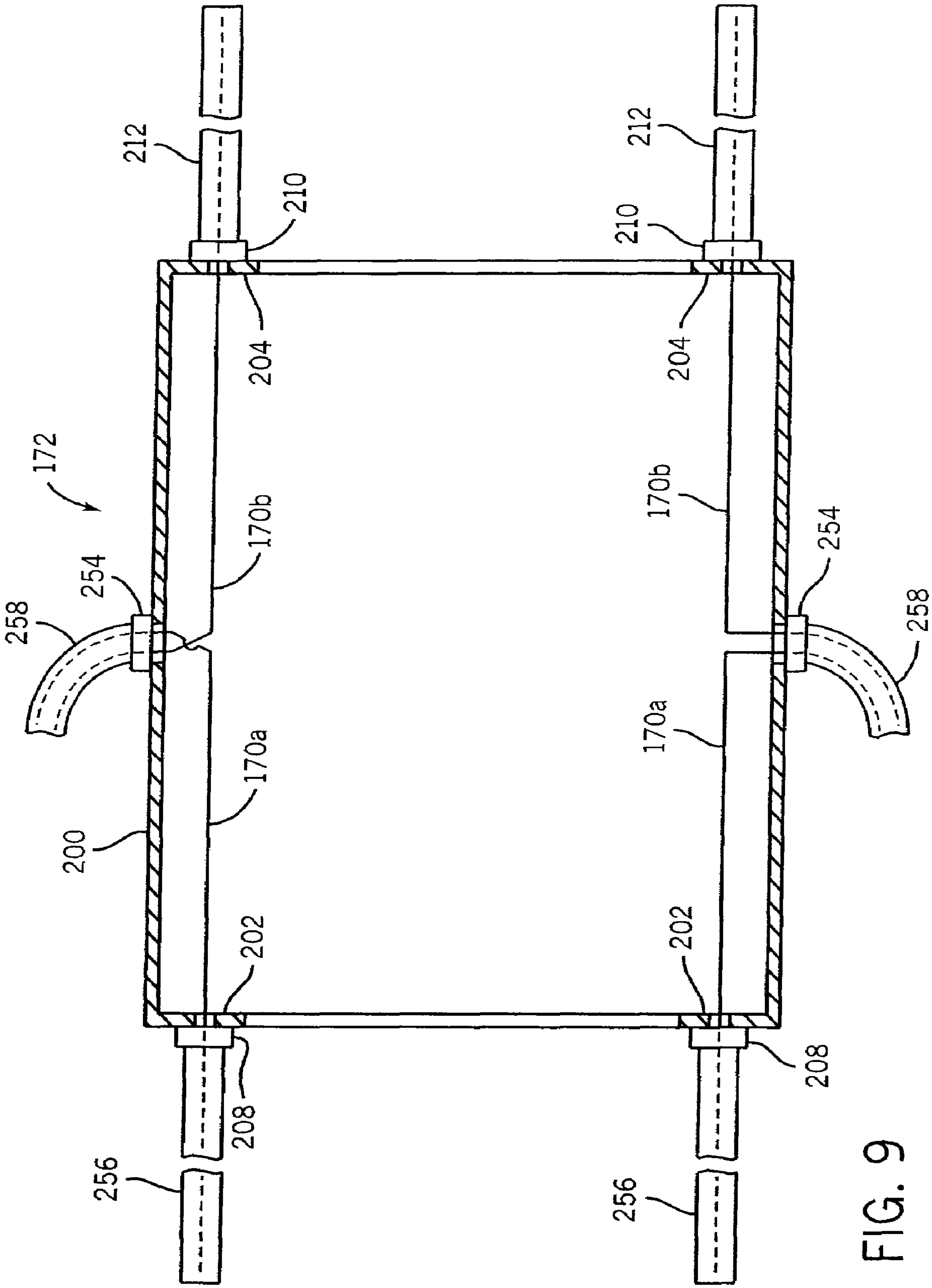


FIG. 9

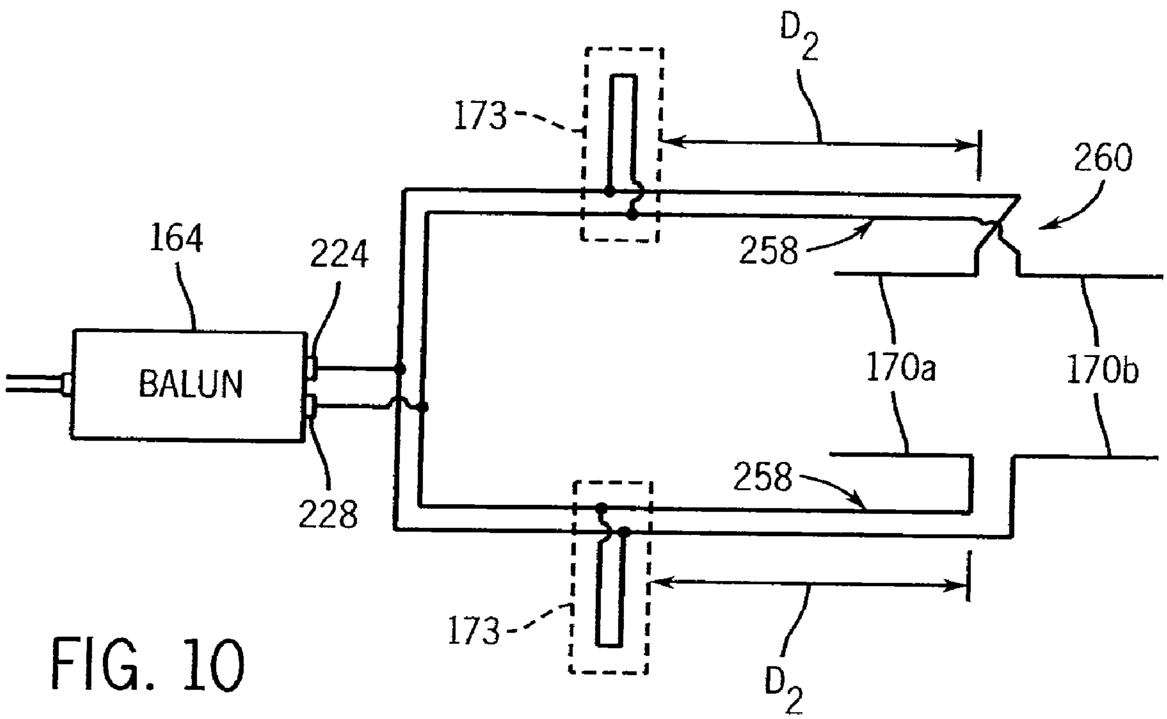
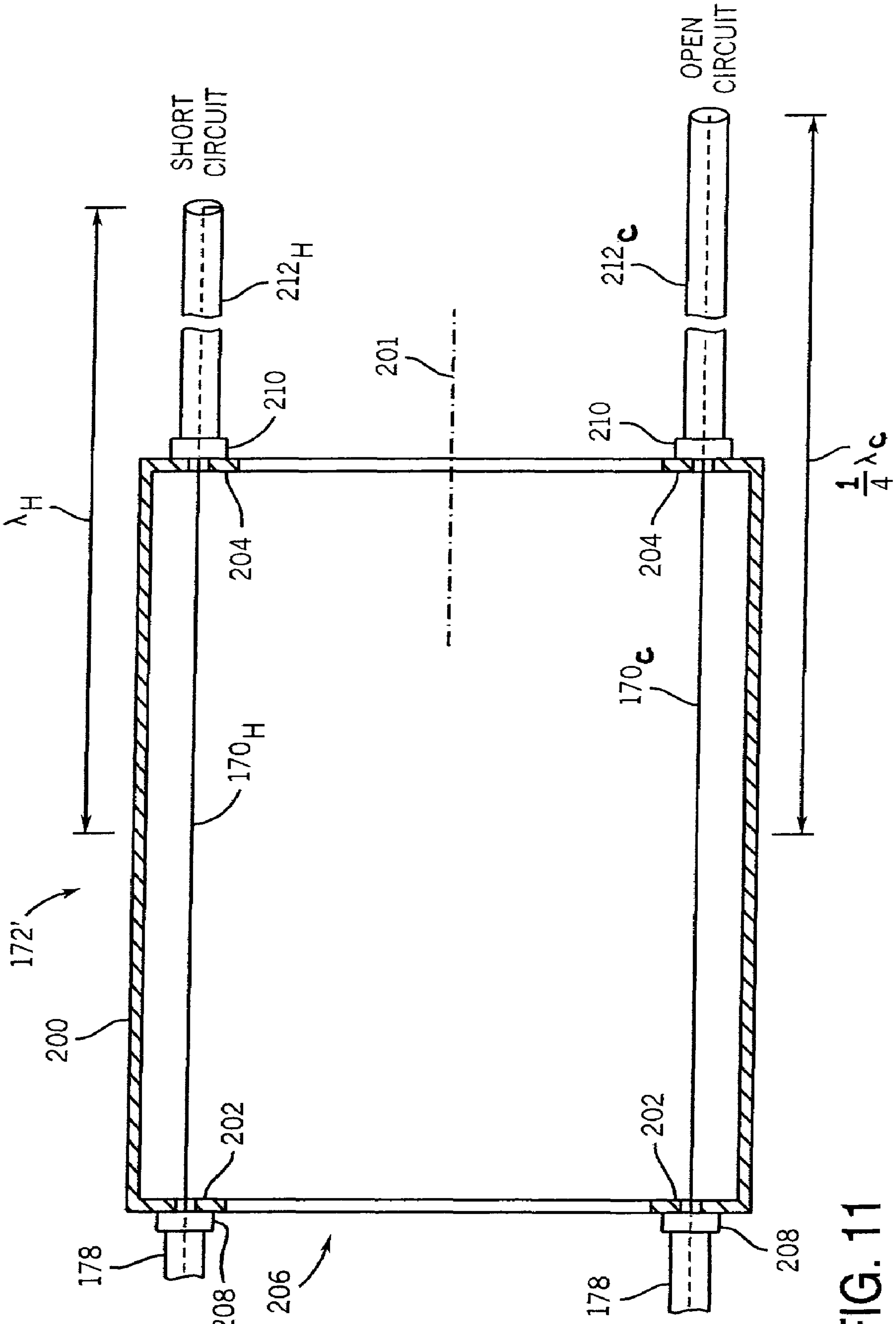


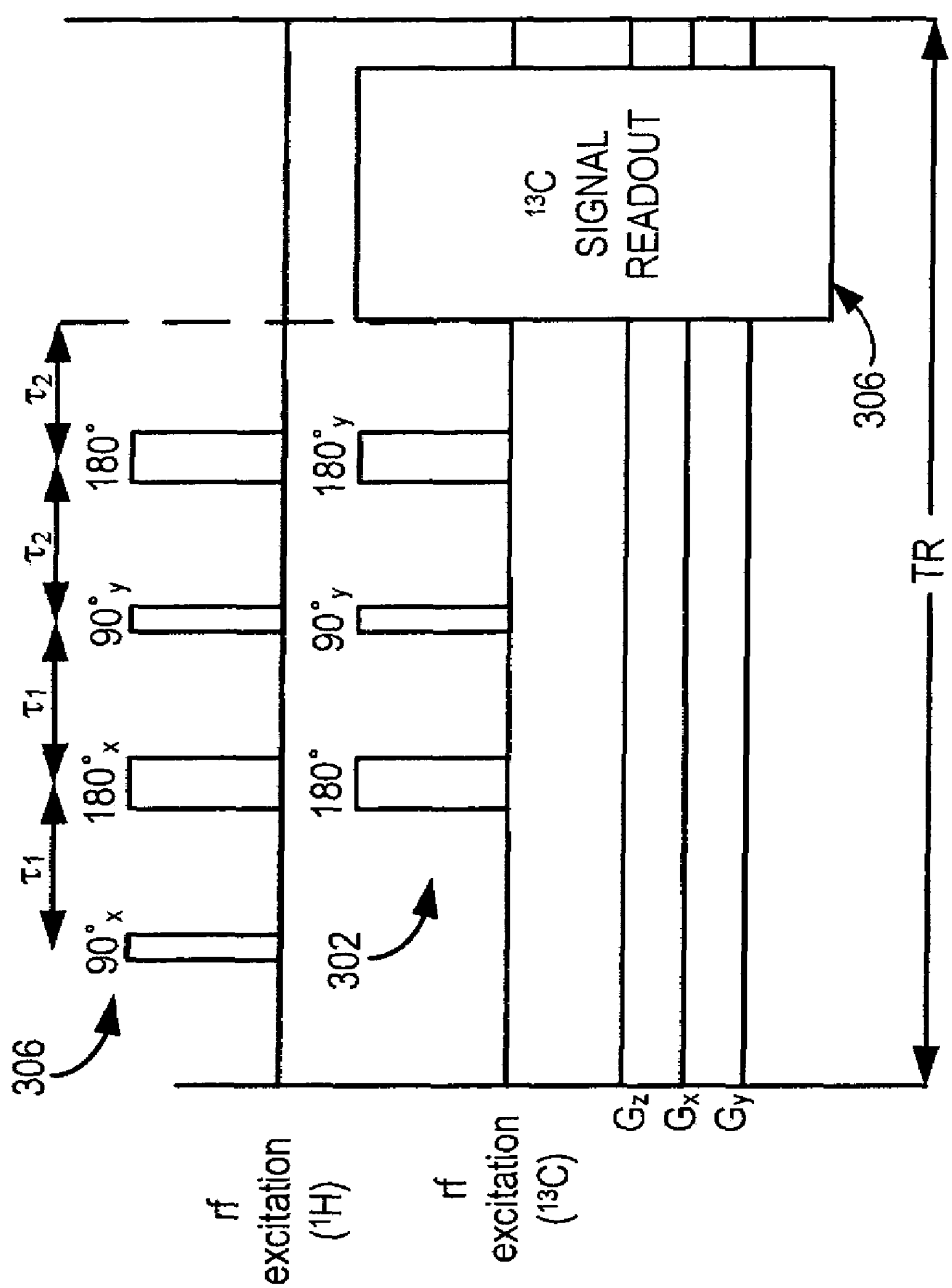
FIG. 10



SHORT CIRCUIT
OPEN CIRCUIT
212_H
212_C
201
210
210
λ_H
204
204
$\frac{1}{4}\lambda_C$
170_C
170_H
172'
200
202
202
178
208
206
178
208

FIG. 11

FIG. 12



rf excitation (^{1}H)
rf excitation (^{13}C)
G_z
G_x
G_y
306
90°$_x$
τ_1
180°$_x$
τ_1
90°$_y$
τ_2
180°
τ_2
302
180°
90°$_y$
180°$_y$
^{13}C SIGNAL READOUT
306
TR

RF COIL ASSEMBLY AND METHOD FOR PRACTICING MAGNETIZATION TRANSFER ON MAGNETIC RESONANCE IMAGING AND SPECTROSCOPY SYSTEMS

CROSS REFERENCE TO RELATED APPLICATIONS

This application is a continuation-in-part of U.S. patent application Ser. No. 11/726,643, filed Mar. 22, 2007, and titled "RF Coil Assembly for Magnetic Resonance Imaging and Spectroscopy Systems", which is hereby incorporated by reference.

STATEMENT REGARDING FEDERALLY FUNDED RESEARCH

This invention was made with United States government support awarded by the following agency: NIH grants: HL069116 and HL066488. The United States has certain rights in this invention.

FIELD OF THE INVENTION

The field of the invention is nuclear magnetic resonance imaging (MRI) and spectroscopy (MRS) methods and systems. More particularly, the invention relates to coils used to apply radio frequency (RF) magnetic fields in magnetization transfer scans and to receive NMR signals produced in response.

BACKGROUND OF THE INVENTION

When a substance such as human tissue is subjected to a uniform magnetic field (polarizing field B_0) applied along the z axis of a Cartesian coordinate system, the individual magnetic moments of the spins in the tissue attempt to align with this polarizing field, but process about it in random order at their characteristic Larmor frequency. If the substance, or tissue, is subjected to a magnetic field (excitation field B_1) which is in the x-y plane and which is near the Larmor frequency, the net aligned moment, Mz, may be rotated, or "tipped", into the x-y plane to produce a net transverse magnetic moment M_t. A NMR signal is emitted by the excited spins after the excitation signal B_1 is terminated, this signal may be received and processed to form an image or produce a spectrum.

When utilizing these signals to produce images, magnetic field gradients (G_x, G_y and G_z) are employed. Typically, the region to be imaged is scanned by a sequence of measurement cycles in which these gradients vary according to the particular localization method being used. The resulting set of received NMR signals are digitized and processed to reconstruct the image using one of many well known reconstruction techniques.

Radio frequency antennas, or coils are used to produce the excitation field B_1 and other RF magnetic fields in the subject being examined. Such coils are also used to receive the very weak NMR signals that are produced in the subject. Such coils may be so-called "whole body" coils that are large enough to produce a uniform magnetic field for a human subject or, they can be much smaller "local" coils that are designed for specific clinical applications such as head imaging, knee imaging, wrist imaging, etc. Local coils may be either volume coils or surface coils.

The most common whole body coil found in commercial MRI systems is the so-called "birdcage" coil first disclosed in U.S. Pat. Nos. 4,692,705; 4,694,255; and 4,680,548. A birdcage coil has a pair of circular end rings which are bridged by a plurality (typically 8 to 24) of equi-spaced longitudinal straight segments. In a primary mode, currents in the straight segments are sinusoidally distributed which results in good B_1 field uniformity across the axis of the coil. Birdcage coils are the structure of choice in horizontal field MRI systems because they produce a homogeneous magnetic B_1 field in the bore of the magnet. When properly designed and constructed, they have a high SNR which enables them to pick up the small NMR signals emanating from the subject under examination.

The birdcage coil is tuned by proper selection of capacitors which are distributed along the lengths of the straight segments, distributed around each end ring or both. Matching and tuning are commonly achieved by connecting variable capacitors in an "L" configuration at the drive ports. Birdcage coils are typically driven at one, two, or more recently, four ports. Multi-port drive, where each drive source is appropriately phased, ensures uniform, circularly polarized B_1 fields in the imaging volume at B_0 field strengths of 1.5 T or less. Efforts to improve the tunability of birdcage coils either provide fewer capacitor adjustments that distort the homogeneity of the B1 field or provide expensive and complex tuning structures such as those described in U.S. Pat. Nos. 6,396,271 and 6,236,206.

There are a number of clinical applications where MR images are acquired at different Larmor frequencies. Hydrogen (H^1) is the spin species of choice for most MR imaging applications, but other paramagnetic spin species such as phosphorus (^{31}P), fluorine (^{19}F), carbon (^{13}C), sodium (^{23}Na), helium (^{3}He) and xenon (^{129}Xe) are also employed. Most of these alternative spin species are of interest in MR spectroscopy, but the use of helium for imaging the lung and carbon-13 metabolites in cancer, for example, have significant clinical potential. As indicated above, the birdcage coil is difficult to tune at more than one Larmor frequency and the substantial change in Larmor frequency required to examine these alternative spin species is not practical.

Multinuclear excitation and reception coils have been proposed. In U.S. Pat. No. 4,799,016 for example, two birdcage coils are formed on one cylindrical substrate, with one coil tuned to hydrogen (^{1}H) and the other tuned to phosphorus (^{13}P). To reduce interaction between the coils, the fields they produce are offset 90° in phase. In U.S. Pat. No. 5,990,681 an RF coil is described which has an adjustment end ring provided on the end of a birdcage coil, wherein the ring can be rotated to change its Larmor frequency. An important limitation of prior multinuclear coils is that they consist of multimodal resonant structures such as birdcage or TEM volume resonators. If one of the resonant modes corresponding to the Larmor frequency of the first nucleus coincides with the fundamental resonant mode corresponding to the Larmor frequency of the second nucleus, the isolation between the two components of the multi-nuclear coil degrades, and the two components of the coil cannot be operated simultaneously. In addition, poor isolation tends to degrade efficiency for each component of the coil in question. In practice, this means that when an image of a subject is acquired at the Larmor frequency of one nucleus, a subsequent scan must be performed if an image is to be obtained at the Larmor frequency of the second nucleus. During the time interval between scans, subject motion may occur, making the co-registration of the two scans difficult. It is therefore desirable to design multi-nuclear coils wherein the component coils are not multi-modal in nature, and the component coils have good electrical isolation and nearly identical spatial profiles.

The in vivo MRS of nuclei other than ^1H provides valuable information about metabolism, and the study of intermediary metabolism of biomolecules provides insight into disease processes. A ^{13}C contrast agent or a ^{31}P contrast agent, for example, may be administered and an MRS acquisition performed to indicate where these agents are used in the subject under examination. Since the MRS images do not reveal the anatomic structures of the subject, it is common practice to also acquire a conventional ^1H image and overlay the MRS image to reveal where in the anatomy the MRS signals are emanating.

The MR signal produced by spin species such as ^{13}C is much lower than that obtained for ^1H and the SNR of the resulting MRS image is low. The SNR may be expressed as:

$$SNR \propto \gamma PC = \gamma^2 B_0 C$$

where γ is the gyromagnetic ratio of the nuclei in question, P is the polarization and C is the concentration of the signal generating nuclei. At body temperature the polarization (P) of ^{13}C is only about one-fourth that of ^1H and its concentration C is also much lower. To overcome this SNR disadvantage, methods have been developed as described by Ardenkjaer-Larsen et al., "Increase in Signal-to-Noise Ratio of >10,000 Times in Liquid-State NMR," PNAS, Sep. 2, 2003, Vol. 100, No. 18, to hyperpolarize the ^{13}C nuclei prior to administration to the subject. Such hyperpolarization can significantly increase the SNR of the MRS image, however, the half life of the hyperpolarized ^{13}C is only 7 to 40 seconds. This requires prompt scanning after administration of the ^{13}C contrast agent.

Another difficulty in acquiring ^{13}C MR signals is that the signals are split due to J-coupling with ^1H spins. This J-coupling reduces sensitivity and spectral resolution. However, the J-coupling can be used to advantage if the ^1H spins are saturated by application of RF energy at their Larmor frequency over a bandwidth of approximately 5 ppm. Through the Nuclear Overhauser Effect (NOE), not only is the split up of the ^{13}C MR signal corrected to increase SNR, but the magnetization M_z of ^{13}C is increased by the transfer of magnetization from ^1H due to their coupling. The trick is to saturate ^1H spins at their Larmor frequency and both excite and readout MR signals at the ^{13}C Larmor frequency during the same scan. Inadequate isolation between resonant modes prevents simultaneous operation of conventional multi-nuclear coil designs, which decreases the efficiency of spin exchange in the NOE experiment, particularly at high field strengths (e.g. 3 Tesla) where SAR limitations also occur. Moreover it is often the case that transmit coils for saturating ^1H spins have a very different spatial sensitivity than the receive coils used for reading out the ^{13}C signal. This leads to spatially variable saturation and magnetization transfer that is undesirable for quantitative imaging applications.

SUMMARY OF THE INVENTION

The present invention is an MRI system which acquires an MRS image of one spin species using magnetization transfer from another spin species having a different Larmor frequency. More particularly, the MRI system includes a resonator having a cylindrical shield formed around a central axis and having plurality of pairs of opposing conductive legs arranged symmetrically around the central axis with the pairs of opposing conductive legs being divided into a first set and a second set, each conductive leg extending from one end of the cylindrical shield to another end of the cylindrical shield. Each set includes: means for tuning each conductive leg in the first set to the Larmor frequency of a first spin species, means for tuning each conductive leg in the second set to the Larmor frequency of a second spin species, first drive circuitry connected to each pair of opposing conductive legs in the first set and being operable to establish substantially equal and opposite current flow in opposing conductive legs at the Larmor frequency of the first spin species, second drive circuitry connected to each pair of opposing conductive legs in the second set and being operable to establish substantially equal and opposite current flow in opposing conductive legs at the Larmor frequency of the second spin species; and means for directing the first drive circuit to saturate the first spin species and thereby transfer magnetization to the second spin species and for directing the second drive circuit to excite the second spin species and acquire an MR signal from the same.

A general object of the invention is to enable magnetization transfer from one spin species to another while acquiring an MR signal from the other spin species. This is achieved by using one set of opposing conductive legs and associated drive circuitry to saturate the one spin species at its Larmor frequency and by using the other set of opposing conductive legs and associated drive circuitry to excite the other spin species and acquire the resulting MR signals therefrom. The excitation volumes are the same for both sets of opposing conductive legs and mutual coupling is minimal.

Another aspect of the invention is a coil assembly that can operate at multiple Larmor frequencies simultaneously during a magnetization transfer acquisition. This multi-nuclear capability is possible by providing multiple resonators having separate pairs of conductive legs within the same cylindrical shield and enclosing the same imaging volume. Separate drive circuits are provided for the separate pairs of conductive legs that are operated at different Larmor frequencies. The terminal susceptance elements associated with a pair of conductive legs are optimized for the Larmor frequency of one nucleus while also effecting a voltage antinode, or high impedance at the Larmor frequency of the other nucleus. Thus, in selected cases of dual nuclear capability, the two channels can be well isolated from each other and enable them to be used simultaneously.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. **1** is a block diagram of an MRI system which employs the present invention;

FIG. **2** is a block diagram of the RF system that forms part of the MRI system of FIG. **1**;

FIG. **3** is a block diagram of a first preferred embodiment of the coil assembly that forms part of FIG. **2**;

FIG. **4** is a pictorial representation of an end view of the RF resonator of FIG. **7**;

FIG. **5** is a block diagram of the drive circuitry for one of the two Larmor frequencies in the assembly of FIG. **3**;

FIG. **6** is a pictorial representation of a first preferred application of the RF resonator for head and brain imaging that forms part of the coil assembly of FIG. **3**;

FIG. **7** is a pictorial representation of a tune and match circuit that forms part of the assembly of FIG. **3**;

FIG. **8** is a pictorial end view of the conductive legs driven at one Larmor frequency by the drive circuitry of FIG. **5**;

FIG. **9** is pictorial representation of a second preferred RF resonator for body imaging applications that forms part of the coil assembly of FIG. **3**;

FIG. **10** is a schematic representation of the changes required to drive the RF resonator of FIG. **9**;

FIG. **11** is a pictorial view of the RF resonator that forms part of the assembly of FIG. **3**; and

FIG. **12** is a graphic representation of a preferred pulse sequence used to direct the operation of the coil assembly of FIG. **3**.

DESCRIPTION OF THE PREFERRED EMBODIMENT

Referring particularly to FIG. **1**, the preferred embodiment of the invention is employed in an MRI system. The MRI system includes a workstation **10** having a display **12** and a keyboard **14**. The workstation **10** includes a processor **16** which is a commercially available programmable machine running a commercially available operating system. The workstation **10** provides the operator interface which enables scan prescriptions to be entered into the MRI system.

The workstation **10** is coupled to four servers: a pulse sequence server **18**; a data acquisition server **20**; a data processing server **22**, and a data store server **23**. In the preferred embodiment the data store server **23** is performed by the workstation processor **16** and associated disc drive interface circuitry. The remaining three servers **18**, **20** and **22** are performed by separate processors mounted in a single enclosure and interconnected using a 64-bit backplane bus. The pulse sequence server **18** employs a commercially available microprocessor and a commercially available quad communication controller. The data acquisition server **20** and data processing server **22** both employ the same commercially available microprocessor and the data processing server **22** further includes one or more array processors based on commercially available parallel vector processors.

The workstation **10** and each processor for the servers **18**, **20** and **22** are connected to a serial communications network. This serial network conveys data that is downloaded to the servers **18**, **20** and **22** from the workstation **10** and it conveys tag data that is communicated between the servers and between the workstation and the servers. In addition, a high speed data link is provided between the data processing server **22** and the workstation **10** in order to convey image data to the data store server **23**.

The pulse sequence server **18** functions in response to program elements downloaded from the workstation **10** to operate a gradient system **24** and an RF system **26**. Gradient waveforms necessary to perform the prescribed scan are produced and applied to the gradient system **24** which excites gradient coils in an assembly **28** to produce the magnetic field gradients G_x, G_y and G_z used for position encoding NMR signals. The gradient coil assembly **28** forms part of a magnet assembly **30** which includes a polarizing magnet **32** and a whole-body RF coil **34** made in accordance with the teachings of the present invention.

RF excitation waveforms are applied to the RF coil **34** by the RF system **26** to perform the prescribed magnetic resonance pulse sequence. Responsive NMR signals detected by the RF coil **34** are received by the RF system **26**, amplified, demodulated, filtered and digitized under direction of commands produced by the pulse sequence server **18**. The RF system **26** includes an RF transmitter for producing a wide variety of RF pulses used in MR pulse sequences. The RF transmitter is responsive to the scan prescription and direction from the pulse sequence server **18** to produce RF pulses of the desired frequency, phase and pulse amplitude waveform. The generated RF pulses may be applied to the whole body RF coil **34** or to one or more local coils or coil arrays.

The RF system **26** also includes one or more RF receiver channels. Each RF receiver channel includes an RF amplifier that amplifies the NMR signal received by the RF coil to which it is connected and a quadrature detector which detects and digitizes the I and Q quadrature components of the received NMR signal. The magnitude of the received NMR signal may thus be determined at any sampled point by the square root of the sum of the squares of the I and Q components:

$$M=\sqrt{I^2+Q^2},$$

and the phase of the received NMR signal may also be determined:

$$\phi=\tan^{-1}Q/I.$$

The pulse sequence server **18** also optionally receives patient data from a physiological acquisition controller **36**. The controller **36** receives signals from a number of different sensors connected to the patient, such as ECG signals from electrodes or respiratory signals from a bellows. Such signals are typically used by the pulse sequence server **18** to synchronize, or "gate", the performance of the scan with the subject's respiration or heart beat.

The pulse sequence server **18** also connects to a scan room interface circuit **38** which receives signals from various sensors associated with the condition of the patient and the magnet system. It is also through the scan room interface circuit **38** that a patient positioning system **40** receives commands to move the patient to desired positions during the scan.

It should be apparent that the pulse sequence server **18** performs real-time control of MRI system elements during a scan. As a result, it is necessary that its hardware elements be operated with program instructions that are executed in a timely manner by run-time programs. The description components for a scan prescription are downloaded from the workstation **10** in the form of objects. The pulse sequence server **18** contains programs which receive these objects and converts them to objects that are employed by the run-time programs.

The digitized NMR signal samples produced by the RF system **26** are received by the data acquisition server **20**. The data acquisition server **20** operates in response to description components downloaded from the workstation **10** to receive the real-time NMR data and provide buffer storage such that no data is lost by data overrun. In some scans the data acquisition server **20** does little more than pass the acquired NMR data to the data processor server **22**. However, in scans which require information derived from acquired NMR data to control the further performance of the scan, the data acquisition server **20** is programmed to produce such information and convey it to the pulse sequence server **18**. For example, during prescans NMR data is acquired and used to calibrate the pulse sequence performed by the pulse sequence server **18**. Also, navigator signals may be acquired during a scan and used to adjust RF or gradient system operating parameters or to control the view order in which k-space is sampled. And, the data acquisition server **20** may be employed to process NMR signals used to detect the arrival of contrast agent in an MRA scan. In all these examples the data acquisition server **20** acquires NMR data and processes it in real-time to produce information which is used to control the scan.

The data processing server **22** receives NMR data from the data acquisition server **20** and processes it in accordance with description components downloaded from the workstation **10**. Such processing may include, for example: Fourier transformation of raw k-space NMR data to produce two or three-dimensional images; the application of filters to a reconstructed image; the performance of a backprojection image reconstruction of acquired NMR data; the calculation of functional MR images; the calculation of motion or flow images, etc.

Images reconstructed by the data processing server **22** are conveyed back to the workstation **10** where they are stored. Real-time images are stored in a data base memory cache (not shown) from which they may be output to operator display **12** or a display **42** which is located near the magnet assembly **30** for use by attending physicians. Batch mode images or selected real time images are stored in a host database on disc storage **44**. When such images have been reconstructed and transferred to storage, the data processing server **22** notifies the data store server **23** on the workstation **10**. The workstation **10** may be used by an operator to archive the images, produce films, or send the images via a network to other facilities.

Referring particularly to FIG. **2**, the RF system **26** includes two transceivers indicated at **80** and **82**. The transceivers are substantially identical in construction, except the transceiver **80** is tuned to the Larmor frequency of hydrogen protons (^1H) and the second transceiver **82** is tuned to the Larmor frequency of carbon (^{13}C). The following is a discussion of the transceiver **80**, but it also applies to the second transceiver **82**. The base, or carrier, frequency of an RF excitation field is produced by the transceiver **80** under control of a frequency synthesizer **100** which receives a set of digital signals from the pulse sequence server **18**. These digital signals indicate the frequency and phase of the RF carrier signal produced at an output **101**. The RF carrier signal is applied to a modulator and up converter **102** where its amplitude is modulated in response to a signal R(t) also received from the pulse sequence server **18**. The signal R(t) defines the envelope of the RF excitation pulse to be produced and is produced by sequentially reading out a series of stored digital values. These stored digital values may be changed to enable any desired RF pulse envelope to be produced.

The magnitude of the RF excitation pulse produced at output **105** is attenuated by an exciter attenuator circuit **106** which receives a digital command from the pulse sequence server **18**. The attenuated RF excitation pulses are applied to the power amplifier **151** that drives an RF coil assembly **155**, through a transmit/receive (T/R) switch **152**. For a more detailed description of this transmitter section reference is made to U.S. Pat. No. 4,952,877 which is incorporated herein by reference.

Referring still to FIG. **2**, the NMR signal produced by the subject is picked up by the coil assembly **155** and applied through the transmit/receive switch **152** to a receiver channel. Although FIG. **2** illustrates a single receiver channel, in other embodiments, the transceiver **80** can include multiple receiver channels. The NMR signals are applied to a receiver preamplifier and attenuator **107** which amplifies the NMR signal by an amount determined by a digital attenuation signal received from the pulse sequence server **18**. The received NMR signal is at or around the Larmor frequency, and this high frequency signal is down converted in a two step process by a down converter **108** which first mixes the NMR signal with the carrier signal on line **101** and then mixes the resulting difference signal with a reference signal on line **104**. The down converted NMR signal is applied to the input of an analog-to-digital (A/D) converter **109** which samples and digitizes the analog signal and applies it to a digital detector and signal processor **110** which produces 16-bit in-phase (I) values and 16-bit quadrature (Q) values corresponding to the received signal. The resulting stream of digitized I and Q values of the received signal are output to the data acquisition server **20**. The reference signal as well as the sampling signal applied to the A/D converter **109** are produced by a reference frequency generator **103**. For a more detailed description of the receiver, reference is made to U.S. Pat. No. 4,992,736 which is incorporated herein by reference.

The transmit/receive switch **152** is used when the RF coil assembly is employed to both produce the uniform B_1 field and receive the resulting NMR signals. It is operated by the pulse sequence server **18** to switch to the transmitter during parts of the pulse sequence in which RF fields are to be produced at the ^1H Larmor frequency, and to switch connection to the receiver when NMR data is to be acquired. The connecting line **153**A between the transmit/receive switch **152** and the coil assembly **155** is preferably a 50 ohm coaxial cable such as RG-213 Mil Spec or Andrews FSJ-50.

As indicated above, the second transceiver **82** is substantially the same as the transceiver **80** except it operates at the Larmor frequency of ^{13}C. it can produce an RF excitation signal that is applied to the coil assembly **155** through a cable **153**B, and it can receive NMR signals from the coil assembly **155** through cable **153**B. The received NMR signals are down converted, digitized and output to the digital detector and processor **110** as described above. An important aspect of this system is that the two transceivers **80** and **82** can be operated simultaneously to excite and receive NMR signals from both ^1H and ^{13}C spins.

The RF coil assembly **155** is shown generally in FIGS. **3** and **4** and it includes two separate channels f_1 and f_2 which connect to the respective transceivers **80** and **82**. Before describing this multinuclear embodiment of the coil assembly **155**, a more detailed description of each channel f_1 or f_2 will be made first with reference to FIGS. **5** and **6**.

Referring to FIGS. **5** and **6**, each channel f_1 and f_2 in the coil assembly **155** includes a phase combiner/splitter **157**, current balun pairs **165**, tune and match circuits **173**, an RF resonator or coil **172**, and terminal susceptance elements such as coaxial termination stubs **212**. The resonator **172** includes a plurality of conductive legs **170** within a cylindrical shield **200**. In brief, pairs of conductive legs **170** operate as balanced transmission lines terminated by terminal susceptance elements such as coaxial stubs **212** and standing waves are established on each conductive leg **170** within the cylindrical shield by proper selection of the terminal susceptance elements. Preferably, the resonator **172** operates with quadrature phasing on both transmit and receive which is provided by the circuit **157** and current balun pair **165**.

The phase combiner/splitter **157** connects to the coaxial line **153** from the transceiver and produces two equal RF currents on preferably 50 ohm coaxial lines **159** and **161** which differ in phase by 90°. In its simplest form the phase combiner/splitter **157** is a 50 ohm T connector with the common connection made to line **153** and each of the T arms connected to one of the lines **159** or **161** through a matched transmission line. One of the transmission line sections is one quarter wavelength longer than the other section to impart a 90° phase difference between currents on lines **159** and **161**. With this particular embodiment, the circuit **157** must be tuned to the particular Larmor frequency being employed. In the alternative, other circuits are well known in the art which can split an RF signal applied through line **153** to quadrature RF signals that are produced on lines **159** and **161**, or which can combine quadrature signals applied through lines **159** and **161** into a single combined RF signal on line **153**. Phase combiner/splitter circuits that can operate at many frequencies are described, for example, in U.S. Pat. No. 5,296,814 and U.S. Pat. No. 7,019,527 which are incorporated herein by reference.

A current balun pair **165** is provided with each balun individually driven by a respective quadrature signal on line **159** or line **161**. While in most cases it is desirable that the quadrature signals on lines **159** and **161** be exactly 90° out of phase with each other, there are instances when adjustments away from exact quadrature are desirable. This can easily be achieved using different lengths of matched transmission line segments, or by using digital phase control.

The I and Q quadrature RF signals on coaxial cable lines **159** and **161** are each coupled to a balanced load by a respective current balun **164** of current balun pair **165**. As is well known in the art, a coaxial cable is an unbalanced feedline and currents can flow on the outer surface of its shield which cause an unbalanced current flow. As will be described below, the balanced load in this embodiment can be viewed as a ½ wave dipole and the purpose of each balun **164** is to insure that the current flow in one arm of the dipole is always substantially equal and opposite to the current flow in the contralateral dipole arm. A balun that fulfills this objective is a "current" balun that maintains the balanced currents in the dipole arms even though the impedances of the two dipole arms may be different. The current balun is important in this application because in MRI it is the current, not the voltage that produces the magnetic field and it is the magnetic field rather than the electric field that is important. Also, whereas the dipole can be constructed with symmetric loads on each dipole arm, in this case the dipole arms are part of a resonator structure described below that receives a subject to be examined. When placed in the resonator, the subject loading often becomes unequal on the two dipole arms. By using the current balun **164**, current balance is maintained despite such asymmetric loading and the homogeneity of the RF magnetic field in the resonator is maintained. The balanced, phase conjugate current flow at the output of the current balun is important for proper operation of the RF resonator **172**. In a preferred embodiment, the common mode rejection of the current balun **164** should be sufficient to closely match the currents in each conductive arm of a conjugate pair. Current baluns are well known devices and preferred embodiments are described in detail in FIGS. **4***a-c* of the above-cited parent application.

Referring again to FIG. **5**, each of the four resonator-side connectors on the current balun pair **165** is connected to a respective conductive leg **170** in the resonator **172** through a respective tune and match circuit **173**. A feedline such as coaxial cable **176** is used to connect a balun terminal **167** to its respective tune and match circuit **173**. Similarly, feedlines such as coaxial cables **178** are used to connect the tune and match circuits to the conductive legs **170** in the resonator **172**.

As shown more particularly in FIG. **7**, one embodiment of the tune and match circuit **173** receives coaxial cables **176** at a characteristic impedance of 50 ohms. The two pieces of coaxial cable **176** extending from a current balun **164** together form a balanced, shielded line that connects conductive housing or box **174**. The coaxial cables **178** extending from tune and match circuit pair housing **174** also have a characteristic impedance of 50 ohms and each has a length selected to present in combination with the tune and match circuit **173** a purely resistive load of 25 ohms at respective match points **181**. Each tune and match circuit operates to null the susceptance component of the admittance at match point **181** (equivalent to nulling the reactive component of the impedance at this point). This purely resistive load of 25 ohms is also desired at output terminals **167** of the current balun **164**, and can be achieved by properly selecting the lengths of the pieces of coaxial cable **176**. In particular, the 25 ohm resistance of the resonator as seen at match point **181** repeats at each of the two terminals **167** on the balanced end of the balun provided that the two lengths of 50 ohm coaxial cable **176** forming the balanced line are approximately λ/2 in length (where λ corresponds to a desired Larmor frequency) or some integral multiple thereof. Since loads across the output side of the balun are in series, the two 25+j0 impedances at output terminals **167** add to 50+j0, which is seen looking into the input terminal (unbalanced port) of each balun **164**.

Referring particularly to FIG. **6**, the first preferred embodiment of the resonator **172** is in the form of a local coil, or head coil. It includes a circular cylindrical shield **200** which extends radially inward toward a central axis **201** at each of its ends to form respective annular shaped mounting rings **202** and **204**. The shield **200** includes an insulating supporting substrate such as a Lucite™ tube and a double foil Faraday copper shield on opposing surfaces of the substrate as described in U.S. Pat. No. 5,680,046 which is incorporated herein by reference. The mounting rings can also be formed from Lucite™ material or other suitable insulating material. In this embodiment the shield **200** is sized to receive the head of a subject to be examined in a circular opening **206** defined by the inner edge of the mounting ring **202**.

The mounting ring **202** provides a surface to which four connectors **208** may be attached for receiving the cables **178** from the drive circuitry described above. Note that the mounting rings **202** and **204** are not required, and in other embodiments a portion or all the connectors can be provided on the shield **200** such that the feed lines form right angles with the conductive legs and/or the terminal susceptance elements form right angles with the conductive legs. As will be described in more detail below, the connectors are preferably equally spaced about the circumference of the mounting ring **202** and symmetry is maintained. The shield on each cable **178** connects to the resonator shield **200** and its center conductor passes through an opening in the mounting ring **202** and connects to one end of a conductive leg **170**. In essence, the resonator shield **200** is an extension of the cable shield which is expanded radially to receive the subject to be examined.

The multiple conductive legs **170** extend through the cavity defined by the shield **200** parallel to a central axis **201** and attach to corresponding connectors **210** attached to the mounting ring **204** at the opposite end of the resonator **172**.

The conductive legs **170** are constructed of a very low resistance material which is formed very thin to avoid eddy currents induced by changing gradient coil fields. Highly polished copper or polished silvered copper is preferred and the legs are shaped as tubes or ribbons to provide mechanical rigidity. For example, conductive copper tape placed over a cylindrical form of Lucite™, polyethylene, or other suitable material offers a convenient and economical way to form an array of conductive legs. Each conductive leg **170** is spaced radially inward from the cylindrical or Faraday shield **200**. The distance of each conductive leg **170** from the adjacent Faraday shield **200** is adjusted to optimize the homogeneity of the RF magnetic field within the resonator central cavity where the subject to be examined resides. The distance will be a function of the diameter of the resonator, but it is not a function of the resonant frequency to which the conductive leg **170** is tuned. Thus, the conductive legs **170** can be tuned to different Larmor frequencies with no mechanical changes within the resonator shield **200**.

The connectors **210** at the back end of the resonator **172** support terminal impedance elements such as termination stubs connected in series with the conductive elements **212** (of coaxial cable) in such a manner that a homogeneous magnetic field is produced within the resonator cavity. Each terminal impedance element is chosen to give a current maximum and a voltage minimum at the midpoint of each conductive leg **170**. In this embodiment, the conductive legs **170** and termination stubs **212** function as a continuous transmission line, albeit with a possible impedance discontinuity at the conductive element-stub electrical junction. We designate the characteristic differential impedance of conductive element conjugate pairs and termination stub pairs as Z_0^{cond} and Z_0^{stub} respectively. It should be noted that these impedances are complex quantities, wherein Z=R+jX. Thus, impedance is the vector sum of a resistive component R and a reactive component X.

To effect a virtual short at the midpoint of each conductive leg requires that the differential impedance of the stub pair, R+jX, be equal to that of an open circuit stub pair of length [¼λ−L/2] whose characteristic impedance Z_0^{stub} is identical to Z_0^{cond}. The quality of this short depends on the Q of the stub **212**: a high Q will effect a virtual short of very high conductance. A similar analysis holds for short circuited stubs. The short circuited termination stubs will perform best when the short circuits are made with high quality preferably silver solder joints, which should be polished and free of oxides for the highest possible RF conductance. It is also good practice to keep the conductive legs **170** polished and free of oxides.

If $Z_0^{cond} \neq Z_0^{stub}$, the effective differential impedance of the stub pair must be determined by normalizing the impedance of the stub with respect to Z_0^{cond}, the characteristic admittance of the conductive leg. In the case where $Z_0^{cond} > Z_0^{stub}$, the stub lengths required for a virtual short at each conductive leg midpoint will be shorter than those required in the case where $Z_0^{cond} = Z_0^{stub}$. This case offers practical advantage as shorter termination stubs are more compact.

One may measure Z_0^{cond} from open circuit and short circuit impedance measurements of a conductive leg conjugate pair at a frequency where the length of the transmission line resonator body corresponds to an odd integral multiple of λ/8. This insures that both measurements will be of comparable magnitude, and won't be near the extremes of an instrument's range. From these measurements, an excellent estimate of Z_0^{cond} can be obtained from the equation:

$$Z_0^{cond} = \sqrt{Z_{sc}^{\text{cond}} \times Z_{oc}^{\text{cond}}}$$

The desired virtual short at conductive leg midpoints may be verified by looking into the input terminal of a current balun whose output terminals are connected to both legs of a conjugate pair of the resonator. If the distance between balun input and resonator leg midpoints is λ/2 or an integral multiple thereof, a short circuit will be seen if the normalized impedance of the stub is correct. It should be noted that no pair of capacitors can substitute for a transmission line stub pair connected in series with a conductive element conjugate pair. Matched capacitve L networks might substitute for stubs, but the shunt capacitor of each L network would have to be extremely small to emulate a good open circuit. Such L networks would be vulnerable to the effects of stray reactance. Thus, transmission line stubs are the preferred embodiment of terminal impedance elements.

As shown in FIG. **8**, the four conductive legs **170** for one frequency channel are spaced evenly around the circumference of the resonator **172**. The symmetry required to obtain a homogeneous RF magnetic field within this resonator demands that current flow be equal and opposite to each other on opposite sides of the central axis **201**. This is achieved by symmetrically positioning pairs of opposing conductive legs around the central axis **201**. In the four element resonator, one opposing pair of conductive legs **170** are positioned at 0° and 180°, and a second opposing pair are positioned at 90° and 270°. This is the geometric aspect of the required symmetry. An additional requirement is that the opposing conductive elements conduct equal, but opposite currents. More conductive legs **170** and associated drive circuitry can be added to improve field homogeneity, but in all cases they must be arranged to maintain the above symmetry.

Another embodiment of the RF resonator **172** that may be suitable for use as a whole-body RF coil **34** is shown in FIG. **9**. The shield **200** and mounting rings **202** and **204** are essentially the same as described above, except they are scaled up in size. To maintain a desired homogeneous RF magnetic field over a larger volume, the conductive legs **170** are divided into two equal leg segments **170*a*** and **170*b*** and driven from a midpoint rather than the front end of the resonator. More particularly, a front conductive leg segment **170*a*** connects at a midpoint connector **254** and extends forward through the connector **208** into a termination stub **256**. Similarly, a rear conductive leg segment **170*b*** connects at the midpoint connector **254** and extends rearward through connector **210** and into the termination stub **212**. The termination stubs **256** and **212** are identical and they are tuned as described above to produce maximum current and minimum voltage at the midpoint of the resonator at a Larmor frequency of choice.

The conductive leg segments **170*a*** and **170*b*** connect to the conductors in a twin-lead, low impedance cable **258** that terminates at the midpoint connector **254**. As shown in FIG. **10**, the twin lead cables **258** extend a distance D to a respective tune and match circuit **173** and the two leads therein extend to and connect to the two phase conjugate connectors **224** and **228** on the balun **164**. Each conductive leg segment pair **170*a*** and **170*b*** forms a dipole antenna and to maintain the symmetry described above that is required for a homogeneous magnetic field, the leads to one of the two dipole antennas are switched as indicated at 260 to change its phase 180° with respect to the opposing dipole antenna.

In the first embodiment of the resonator **172** described above there exists a virtual short circuit at the midpoint of each conductive leg **170**. This virtual short point is a voltage node where the electric field is very small and dielectric losses are minimal. The current distribution along the length of each conductive leg **170** is peak at its midpoint and drops off sinusoidally as a function of distance from this midpoint and the wavelength (λ) corresponding to the Larmor frequency. This drop-off condition requires that the axial length of the resonator **172** be limited to less than λ/10 in order to maintain homogeneous magnetic fields and keep electric field magnitudes acceptably low. This is not a problem when the resonator is used at lower polarizing fields (i.e. longer Larmor frequency wavelength) or the resonator is a relatively small local coil.

The embodiment of the resonator **172** illustrated in FIG. **9** relieves this constraint by a factor of two. By driving each conductive leg **170** at its midpoint rather than one end, the length of the resonator is limited to λ/5 rather than λ/10. This results because the end of each conductor at the feed point is precisely ¼ wavelength away from the open circuit stub end, or is an odd integral multiple thereof. Alternatively, if this stub end is short circuited, the wavelength is n*λ/4, where n is an even integer.

The coil assembly **155** in FIG. **2** can be used to perform multinuclear NMR measurements simultaneously. This is achieved by tuning selected pairs of the conductive legs **170** to the Larmor frequency of one spin species and tuning the Larmor frequency of the remaining conductive legs **170** to another spin species. For example, one Larmor frequency may be that of hydrogen protons and the other Larmor frequency may be that of ^3He, ^{13}C, or ^{23}Na.

Referring particularly to FIGS. **3** and **4**, the multinuclear coil assembly **155** for driving both frequency channels includes an eight-element RF double resonator **172'**. The f_1 transceiver **80** couples to one set of drive circuitry indicated at **300** and the other transceiver **82** couples to a second set of drive circuitry indicated at **302**. The drive circuits **300** and **302** are identical to those described above with respective phase combiner/splitters **157**A, **157**B, current balun pairs **165**A, **165**B and tune and match circuit's pairs **174**A, **174**B. For drive circuitry **300**, the distance D_1 between the tune and match circuit pair **174**A and the RF double resonator **172'** is adjusted to the length needed for the wavelength λ_1 corresponding to the Larmor frequency f_1, in the same manner described above. Further, the four coaxial stubs **212**A for the leg segments **170**A driven by the f_1 channel are also adjusted in length. The f_1 tune and match circuit pair **174**A will also be set to null any reactive component at this frequency as described above. Similarly, corresponding adjustments are made for corresponding distance D_2, length of coaxial stubs **212**B, tune and match circuit pair **174**B in the f_2 channel such that they are tuned to the wavelength λ_2 corresponding to the second Larmor frequency f_2.

As shown in FIG. **4**, the four conductive legs **170**A driven by the f_1 channel are positioned at the physical angles 0°, 90°, 180° and 270° around the circumference of the double resonator **172'** and as described above, their associated coaxial stubs **212**A are set to the λ_1 wavelength. The remaining four conductive legs **170**B at physical angles 45°, 135°, 225° and 315° are driven by the f_2 channel and their associated coaxial stubs **212**B are set to the λ_2 wavelength.

Thus the drive circuitry for the f_1 conductive legs **170**A and the f_2 conductive legs **170**B operate exactly as described above for a coil assembly operating at a single Larmor frequency. More specifically, the quadrature rotating field for each frequency f_1 and f_2 is supported by pairs of opposing conductive legs **170**A, **170**B oriented 90° with respect to one another. The conductive legs **170**A are interspersed with the conductive legs **170**B around the mounting ring **204** at 45° increments with respect to one another shown in FIG. **4** without changing the pattern of their excitation. Note that the electrical phase increments will be the same for both the f_1 coil and the f_2 coil, namely 0°, 90°, 180°, 270°. This arrangement would normally be problematic due to the magnetic coupling between adjacent conductive legs **170**A, **170**B at respective frequencies f_1 and f_2 and a severe distortion of the B_1 RF field at each frequency would result for both transmit and receive.

Fortunately, nature presents a solution to this coupling problem for some combinations of NMR active nuclei that enables them to be used together. It may be seen that the ratio of the gyromagnetic ratios of ^{3}He and ^{1}H (proton) are approximately 0.762 as determined by accurate experiment. This ratio is very close to ¾, and allows for excellent decoupling between the two resonators as will be described below. It may also be seen that the ratio of the ^{13}C and ^{1}H gyromagnetic ratios is very close to ¼; this permits the construction of dual frequency ^{13}C-^{1}H resonators with excellent mutual isolation between the two. This approach may also be used for ^{23}Na and ^{1}H.

Referring particularly to FIG. **11**, the double frequency resonator **172'** is shown with two pairs of conductive legs, one pair $\mathbf{170}_H$ tuned to the Larmor frequency of ^{1}H and the other pair $\mathbf{170}_C$ tuned to the Larmor frequency of ^{13}C. The coaxial stub $\mathbf{212}_H$ on the hydrogen conductive leg $\mathbf{170}_H$ is set to produce a current maximum at its midpoint as described above. This can be achieved with a shorted stub and a distance between the shorted end of the stub and the midpoint of conductive legs approximately equal to one wavelength λ_H. This length is chosen because at the Larmor frequency of ^{13}C, this length produces a current minimum for ^{1}H nuclei at the same midpoint. Similarly, the coaxial stub $\mathbf{212}_C$ for the ^{13}C conductive leg $\mathbf{170}_C$ is set to produce a current maximum at the midpoint of the conductive leg $\mathbf{170}_C$. This can be achieved with an open stub and a distance between the open end of the stub on the midpoint of the conductive leg approximately equal to a length of $\frac{1}{4}\lambda_C$. This length produces a current minimum at the same approximate midpoint for signals at the hydrogen Larmor frequency. A high degree of isolation will be realized because each resonator acts as a stop band filter for the other. This may be understood by considering the load seen by a conductive leg $\mathbf{170}_C$ if it were to couple with a neighboring conductive leg $\mathbf{170}_H$: the coaxial stub $\mathbf{212}_H$ inverts a short at the ^{13}C frequency, so a very high load is seen by the conductive leg $\mathbf{170}_H$ and coupling is minimized. Similarly, a conductive leg $\mathbf{170}_H$ sees a very high impedance on neighboring conductive leg $\mathbf{170}_C$.

This illustrates why termination stubs are used as terminal impedance elements in the double nuclear coil: unlike a capacitor, a stub can "switch" from inductive (+) reactance to capactive (−) reactance with a change in frequency. In this case, a change in frequency approximates a complex inversion of the stub's susceptance at the generator end of the stub. In other words, the stub goes from Z to approximately (−1/Z). Briefly, the best isolation comes when the conductive leg is a near short at its Larmor frequency and a near open circuit at the other Larmor frequency. This means that the respective impedances measured at the midpoint of the conductive leg at the two Larmor frequencies will be as far apart as possible on the real axis of the Smith Chart.

Thus, by judiciously tuning the conductive legs **170** they can be rendered sensitive to signals at only one of the two Larmor frequencies being used. Table 2 is a list of possible combinations and the tuning of the stubs.

TABLE 2

Multiple Frequency Coil Table	
^{3}He/^{1}H Dual Coil	¾ wave OC stubs for ^{3}He and one wave SC stubs for ^{1}H
^{13}C/^{1}H Dual Coil	¼ wave OC stubs for ^{13}C and one wave SC stubs for ^{1}H
^{23}Na/^{1}H Dual Coil	¼ wave OC stubs for ^{23}Na and one wave SC stubs for ^{1}H

OC = open circuit
SC = short circuit

To practice the present invention the MRI system operates under the direction of a pulse sequence that directs the operation of the two drive circuits in the above-described coil assembly. Referring particularly to FIG. **12**, the ^{1}H spins are excited at their Larmor frequency by a series of rf pulses indicated generally at **300**, and the ^{13}C spins are excited at their Larmor frequency by a series of rf pulses indicated generally at **302**. Several polarization (magnetization) transfer pulse sequences have been developed using the Nuclear Overhauser Effect (NOE) and are known in the art by acronyms including DEPT, INEPT, reverse INEPT described in detail by Morris and Freeman, *Journal of the American Chemical Society*, 1979; 101:760-762; Farrar '*Introduction to Pulse NMR Spectroscopy*' 1997 The Farragut Press Madison, Wis.; Harris '*Nuclear Magnetic Resonance Spectroscopy A Phisiochemical View*' 1983 Pitman Publishing, Marshfield, Mass.; Sohar '*CRC: Nuclear Magnetic Resonance Spectroscopy Volume* 1' 1983 CRC Press Inc, Boca Raton, Fla.; Goldman '*Quantum Description of High-Resolution NMR in Liquids*' 1988 Oxford University Press Inc, New York. Doddrell D M, *J Magn Reson* 48:323-327. The series of rf pulses **300** are applied through one drive circuit to produce balanced currents in opposing conductive legs tuned to ^1H and the series of rf pulses **302** are applied through the other drive circuit to produce balanced currents in opposing conductive legs tuned to ^{13}C. The abundant transverse ^1H magnetization produced by a 90° rf pulse **304** is transferred to the ^{13}C spin magnetization by the J-coupling, or Nuclear Overhauser Effect, which arises from the dipolar cross-relaxation between these two nuclei. The rf pulses **300** and **302** operate to produce the ^1H transverse magnetization, to transfer the ^1H magnetization to the ^{13}C spin population and then tip that ^{13}C magnetization back to the longitudinal axis. Thus, the rf pulses **300** and **302** serve as a preparatory pulse sequence that produces abundant ^{13}C longitudinal magnetization for an imaging pulse sequence to follow. Although the rf pulses **300** and **302** are at different Larmor frequencies, the isolation of the ^{13}C coil elements and the ^1H coil elements insures optimal performance.

Depending on the application, various chemical shift imaging pulse sequences indicated at **306** are performed after the preparatory sequence. For hyperpolarized ^{13}C applications, this can be a rapid imaging sequence such as a echo planar spectroscopic imaging (EPSI) sequence or the IDEAL method (Reeder J Magn Reson Imaging. 2007 March; 25(3): 644-52). This pulse sequence **306** is played out at the ^{13}C Larmor frequency to read out an MR signal produced by the enhanced ^{13}C magnetization. During the readout of the ^{13}C MR signals, only the opposing conductive legs tuned to ^{13}C are employed. They are employed to excite the enhanced ^{13}C magnetization and then acquire the ^{13}C MR signal that is produced during readout. In the preferred embodiment a chemical shift imaging sequence (Brown, Kincaid, and Ugurbil, 1982, PNAS 79:3523-36.) such as the PRESS, STEAM, EPSI or IDEAL sequences are employed to readout the ^{13}C MR signal. The pulse sequence in FIG. **12** is repeated many times and the imaging gradients G_z, G_x, and G_y are stepped through a sequence of values to spatially encode the acquired MR signals in the well-known manner. An image is then reconstructed from this acquired k-space data using an image reconstruction method appropriate with the selected imaging pulse sequence **306**.

Specific advantages of the described invention for polarization transfer are the following. First, true simultaneous excitation can be performed to provide more efficient spin coupling and exchange. Second, the conductor sets tuned to ^1H and ^{13}C are sensitive to nearly identical volumes and therefore polarization or magnetization transfer is more efficient and consistent over the entire imaging volume. Third, decoupling experiments can be performed more efficiently again because of simultaneous ^1H and ^{13}C transmission and excitation. More efficient coupling at higher field strengths can also help mitigate SAR for these applications. Finally, the coil is fully compatible with conventional ^1H MRI and can therefore obtain a spatially registered anatomic map to relate structure and function in spectroscopic imaging applications.

It is also possible to monitor polarization transfer experiments dynamically using this RF coil design. Such an embodiment would use the INEPT or DEPT pulse sequence combined with simultaneous ^1H and ^{13}C image readout. When using hyperpolarized ^1H or ^{13}C labelled compounds, there may be applications for dynamic imaging of metabolism in cancer and other diseases. Moreover, the short T_1 times of hyperpolarized ^1H and ^{13}C compounds would also result in instances where certain sites on a molecule may have longer T_1 relaxation times but be of less biological interest. In vivo transfer of this polarization to other sites of more biological interest on the molecule but with shorter T_1 times might be efficiently performed using the present invention. This makes hyperpolarized ^{13}C imaging more practical in vivo.

The invention claimed is:

1. A magnetic resonance imaging (MRI) system for acquiring magnetic resonance data from one spin species using magnetization transfer from another spin species having a different Larmor frequency, the combination comprising:

a resonator having a cylindrical shield formed around a central axis and having plurality of pairs of opposing conductive legs arranged symmetrically around the central axis with the pairs of opposing conductive legs being divided into a first set and a second set, each conductive leg extending from one end of the cylindrical shield to another end of the cylindrical shield;

means for tuning each conductive leg in the first set to the Larmor frequency of the first spin species;

means for tuning each conductive leg in the second set to the Larmor frequency of the second spin species;

first drive circuitry connected to each pair of opposing conductive legs in the first set and being operable to establish substantially equal and opposite current flow in opposing conductive legs at the Larmor frequency of the first spin species;

second drive circuitry connected to each pair of opposing conductive legs in the second set and being operable to establish substantially equal and opposite current flow in opposing conductive legs at the Larmor frequency of the second spin species; and

means for directing the first drive circuit to excite the first spin species and thereby transfer magnetization to the second spin species and for directing the second drive circuit to excite the second spin species and acquire a magnetic resonance signal from the same.

2. The system as recited in claim **1**, wherein the first and second means for tuning include termination stubs, each termination stub is connected to one end of a respective conductive leg in series.

3. The system as recited in claim **2**, wherein each of the termination stubs for the first means for tuning has a length that combined with a half length of a respective conductive leg is approximately equal to $n_1\lambda_1/4$, where λ_1 is a wavelength corresponding to the first Larmor frequency, and n_1 is an odd integer if the termination stub is open circuited, and n_1 is an even integer if the termination stub is short circuited, and further wherein each of the termination stubs for the second means for tuning has a length that combined with a half length of a respective conductive leg is approximately equal to $n_2\lambda_2/4$, where λ_2 is a wavelength corresponding to the second Larmor frequency, and n_2 is an odd integer if the termination stub is open circuited, and n_2 is an even integer if the termination stub is short circuited.

4. The system as recited in claim **2**, wherein each of the termination stubs for the first means for tuning are open circuited at a distal end thereof and each has a length that combined with a half length of a respective conductive leg is approximately equal to $n_1\lambda_1/4$, where λ_1 is a wavelength corresponding to the first Larmor frequency and n_1 is an odd integer, and further wherein each of the termination stubs for the second means for tuning are short circuited at a distal end thereof and each has a length that combined with a half length of a respective conductive leg is approximately equal to λ_2, where λ_2 is a wavelength corresponding to the second Larmor frequency.

5. The system as recited in claim 1 in which the first spin species is hydrogen (^{1}H) and the second spin species is carbon (^{13}C).

6. The MRI system as recited in claim 1 in which the means for directing the first and second drive circuits includes a pulse sequencer programmed to perform a prescribed pulse sequence.

7. The MRI system as recited in claim 6 in which the prescribed pulse sequence is a magnetization transverse type pulse sequence.

8. The MRI system as recited in claim 6 in which the prescribed pulse sequence is an INEPT pulse sequence and readout simultaneously acquires images of ^{1}H and ^{13}C nuclei.

9. The MRI system as recited in claim 1 in which the first spin species is hyperpolarized.

10. The MRI system as recited in claim 9 in which the hyperpolarized first spin species is ^{13}C.

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