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(54) **Title**: MULTI-ELEMENT TRANSMIT RF CHAIN WITH LOCAL AUTOMATIC TUNE AND MATCH DEVICE

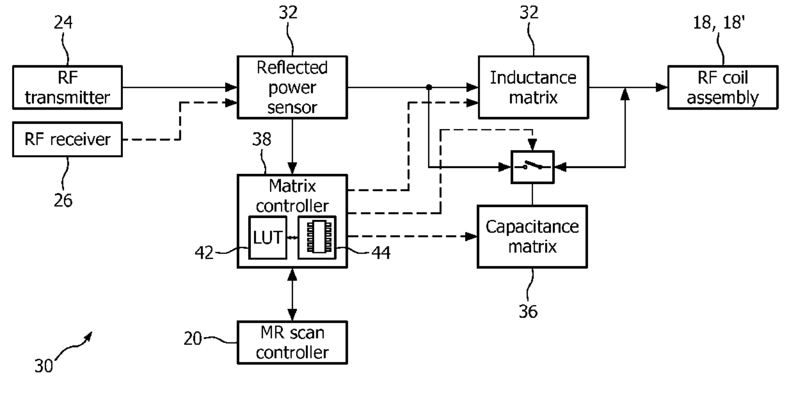

FIG. 3

(57) **Abstract**: An automatic tune and match device (3) and method comprises a reflected power sensor (32) which detects power reflected from a load (18,18') and an LC matching circuit, in series with the load, being programmable to minimize the reflected power. The LC matching circuit includes an inductor matrix (34) in series with the load (18, 18') and a capacitor matrix (36) in parallel with the inductor matrix. A matrix controller (38) configures at least one of the inductor matrix or capacitor matrix based on the detected reflected power to minimize the reflected power.

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MULTI-ELEMENT RF TRANSMIT COIL FOR MRI WITH LOCAL AUTOMATIC TUNE AND MATCH CIRCUIT

DESCRIPTION

The present application relates to high power radiofrequency (RF) impedance matching. It finds particular application to the isolation or impedance matching of RF power amplifiers with multi-element transmit coils in magnetic resonance systems.

Magnetic resonance imaging (MRI) and spectroscopy (MRS) systems are often used for the examination and treatment of patients. By such a system, the nuclear spins of the body tissue to be examined are aligned by a static main magnetic field B_0 and are excited by transverse magnetic fields B_1 oscillating in the radiofrequency band. In imaging, relaxation signals are exposed to gradient magnetic fields to localize the resultant resonance. The relaxation signals are received and reconstructed into a single or multi-dimensional image. In spectroscopy, information about the composition of the tissue is carried in the frequency component of the resonance signals.

Two types of MR systems that are in common use include “open” MR systems (vertical system) and “bore-type” systems. In the former, the patient is introduced into an examination zone which is situated between two magnetic poles connected by a C-shaped unit. The patient is accessible during the examination or treatment from practically all sides. The latter comprises a cylindrical examination space (axial system) into which a patient is introduced.

An RF coil system provides the transmission of RF signals and the reception of resonance signals. In addition to the RF coil system which is permanently built into the imaging apparatus, special purpose coils can be flexibly arranged around or in a specific region to be examined. Special purpose coils are designed to optimize signal-to-noise ratio (SNR), particularly in situations where homogeneous excitation and high sensitivity detection is required. Furthermore, special sequences of RF signals, higher field strengths, high flip angles or real-time sequences can be realized and generated by multi-channel antenna arrangements, and multi-dimensional excitations can be accelerated.

In multi-element transmit coil (multix) systems, each individual coil element is connected to an RF power amplifier. Multix systems can improve B1 magnetic field homogeneity and reduce specific absorption rate (SAR) in patients which permits operation at higher field strengths, e.g. 2 Tesla (T) or higher. Several problems arise from connecting individual coils directly to the RF power amplifier at higher field strengths. The power amplifiers are pre-tuned to selected impedances, e.g. 50 ohms. Matching circuits match the impedance of each coil element to the preselected impedance. However, the patient changes the loading on the coil elements which changes their impedance causing an impedance mismatch. With the impedance mismatch, RF power is reflected back to the power amplifier which wastes power intended to be delivered to the coil element. This results in insufficient insulation between individual coil elements, thus insufficient isolation at the output port of the power amplifier which ultimately yields a non-linear response on the power amplifier.

To address problems with power amplifier isolation, waveguide circulators, or isolators, have been introduced. Circulators are basic three-port non-reciprocal components used to separate incident and reflected waves. When a magnetizing field is created within its ferrite core by insulated conductor windings, a gyromagnetic effect is generated which can be used for circulating a signal from one port to another. The incident signal circulates in only one direction, namely, clockwise or counterclockwise, to reach the next port. If one of the ports is terminated in a matched load, then the circulator acts as an isolator, with high loss in one direction and low loss in the other direction. Therefore, in the reverse direction the ports are isolated from each other and signal propagation is restricted. Magnetic ferrite cores are the most popular material to make passive circulators due to their excellent (RF) performance and lack of moving parts. A desired response occurs within a specific frequency range that can be achieved by modulating the dimensions of the ferrite core and the magnitude of the static magnetic field, i.e. at higher power a larger core is necessary.

High power circulators, such as those used in MR systems, are expensive to design and manufacture. They require large ferrite cores and complicated heat exchange systems that include heat sinks and expensive thermally conductive materials with low dielectric constants to prevent arching. Additional load is required to induce the gyromagnetic effect in larger ferrite cores. Due to the saturation effects of the ferrite core and their intrinsic magnetic nature, the circulators must be position at a distance from the MR main magnet. This forces the RF power amplifiers to be positioned at an even further distance which can increase the already high cost of RF energy and add cabling complexity to the MR room. Furthermore, during operation of a circulator, the reflected RF power heats the ferrite core and leads to unreliable operation which can lead to non-linear of the RF power amplifier and reduce isolation at the ports of the amplifier.

The present application provides a new and improved automatic tune and match apparatus and method which overcomes the above-referenced problems and others.

In accordance with one aspect, an automatic tune and match device comprises a reflected power sensor which detects power reflected from a load and an LC matching circuit, in series with the load, being programmable to minimize the reflected power.

In accordance with another aspect, a method for impedance matching, comprises detecting reflected power from a load programming an LC matching circuit to minimize the reflected power.

One advantage relies in that signal-to-noise ratio (SNR) is increased.

Another advantage relies in that radiofrequency (RF) power requirements are reduced.

Another advantage relies in that system complexity is reduced.

Another advantage relies in that manufacturing costs is reduced.

Another advantage relies in that scan time is reduced.

Still further advantages of the present invention will be appreciated to those of ordinary skill in the art upon reading and understand the following detailed description.

The invention may take form in various components and arrangements of components, and in various steps and arrangements of steps. The drawings are only for purposes of illustrating the preferred embodiments and are not to be construed as limiting the invention.

FIGURE 1 is a diagrammatic illustration of one aspect of a magnetic resonance system with an automatic tune and match unit;

FIGURE 2 is a diagrammatic illustration of another aspect of a magnetic resonance system with an automatic tune and match unit;

FIGURE 3 is a diagrammatic illustration of an automatic tune and match unit;

FIGURES 4A-4C are circuit diagrams of matching circuit configurations of an automatic tune and match unit;

FIGURE 5A is a diagrammatic illustration of a transverse electromagnetic (TEM) coil with an integrated automatic tune and match unit;

FIGURE 5B is a diagrammatic illustration of a loop coil with an integrated automatic tune and match unit;

FIGURE 6A is a diagrammatic illustration in partial of a magnetic resonance system with a recessed gradient coil;

FIGURE 6B is a diagrammatic illustration of a coil element with an integrated automatic tune and match unit;

FIGURE 7 is a diagrammatic illustration of one aspect of a multi-frequency RF coil assembly with an automatic tune and match unit;

FIGURE 8 is a diagrammatic illustration of another aspect of a multi-frequency RF coil assembly with an automatic tune and match unit; and

FIGURE 9 is a timing diagram of an MR sequence and automatic tune match.

With reference to FIGURE 1, a magnetic resonance (MR) imaging system **10** includes a main magnet **12** which generates a temporally uniform B_0 field through an examination region **14**. The main magnet can be an annular or bore-type magnet, a C-shaped open magnet, other designs of open magnets, or the like. Gradient magnetic field coils **16** disposed adjacent the main magnet serve to generate magnetic field gradients along selected axes relative to the B_0 magnetic field for spatially encoding magnetic resonance signals, for producing magnetization-spoiling field gradients, or the like. The magnetic field gradient coil **16** may include coil segments configured to produce magnetic field gradients in three orthogonal directions, typically longitudinal or z, transverse or x, and vertical or y-directions.

A radio-frequency (RF) coil assembly **18**, such as a whole-body radio frequency coil, is disposed adjacent the examination region. The RF coil assembly generates radio frequency B_1 pulses for exciting magnetic resonance in the aligned dipoles of the subject. The radio frequency coil assembly **18** also serves to detect magnetic resonance signals emanating from the imaging region. Optionally, local, surface, head, or in vivo RF coils **18′** are provided in addition to or instead of the whole-body RF coil **18** for more sensitive, localized spatial encoding, excitation, and reception of magnetic resonance signals. In a multi-element RF coil assembly, the RF coil assembly includes a plurality of individual coil elements to improve B_1 homogeneity and reduce specific absorption rate (SAR) in the subject.

To acquire magnetic resonance data of a subject, the subject is placed inside the examination region **14**, preferably at or near an isocenter of the main magnetic field. A scan controller **20** controls a gradient controller **22** which causes the gradient coils to apply the selected magnetic field gradient pulses across the imaging region, as may be appropriate to a selected magnetic resonance imaging or spectroscopy sequence. The scan controller **20** controls an RF transmitter **24** which causes the RF coil assembly to generate magnetic resonance excitation and manipulation B_1 pulses. In a multi-element RF coil assembly, the RF transmitter **24** includes a plurality of transmitters or a single transmitter with a plurality of transmit channels, each transmit channel includes an RF power amplifier operatively connected to a corresponding coil element of the coil assembly. In a single coil design, a single transmit channel includes a single RF power amplifier which generates the excitation and manipulation signals.

The scan controller also controls an RF receiver **26** which is connected to the RF coil assembly to receive the generated magnetic resonance signals therefrom. In a multi-element RF coil assembly, the RF receiver **26** includes a plurality of receivers or a single receiver with a plurality of receive channels, each receive channel includes a pre-amplifier operatively connected to a corresponding coil element of the coil assembly. In a single coil design, a single receive channel includes a single pre-amplifier which amplifies the received magnetic resonance signals.

The scan controller also controls a switching unit **28** which switches the RF coil assembly **18**, **18’** between a transmit mode and a receive mode by selectively coupling the RF coil assembly **18**, **18’** to one of the RF transmitter **24** or RF receiver **26**. In a multi-element RF coil assembly, the switching unit 28 includes a plurality of switches, each switch selectively switches an individual coil element to one of the corresponding RF transmit channels of the RF transmitter **24** or one of the corresponding RF receive channels of the RF receiver **26**.

Typically, prior to assembling an MR system, the individual coil elements of the RF coil assembly are tuned to match the output port of a corresponding RF power amplifier. However, at high field strengths, e.g. 2 Tesla (T) or higher, the subject changes the loading on the each individual coil elements which changes their impedance. An impedance mismatch can cause part or the entire signal generated by the RF power amplifier to be reflected back to the RF transmitter **24** wasting costly RF energy, possibly distorting the incident signal, and possibly damaging the RF transmitter.

In a first aspect, an automatic tune and match unit (ATMU) **30** is disposed between RF transmitter **24** and the switching unit **28**. In a multi-element RF coil assembly, the ATMU **30** includes a plurality of ATMUs, each ATMU is disposed between a corresponding RF transmit channel of the RF transmitter **24** and a corresponding switch of the switching unit **28**. When the MR system **10** is in a transmit mode, the switching unit **28** selectively couples the RF transmitter **24** to the RF coil assembly **18**, **18'**. Each ATMU **30** detects an impedance mismatch between each coil element of the RF coil assembly **18**, **18'** and each corresponding RF power amplifier of the RF transmitter **24**. The scanner controller **20** controls the ATMUs to compensate for the detected impedance mismatch eliminating part or the entire reflected signal.

With reference to FIGURE 2, in a second aspect, an ATMU **30** is disposed between the switching unit **28** and the RF coil assembly **18**, **18'**. When the MR system **10** is in a transmit mode, the switching unit **28** selectively couples the RF transmitter **24** to the RF coil assembly **18**, **18'** and the ATMU **30** detects and compensates for impedance mismatches between the RF coil assembly **18**, **18'** and the RF transmitter **24** or, in the multi-element RF coil assembly example, between each coil element of the RF coil assembly **18**, **18'** and each corresponding RF power amplifier of the RF transmitter **24**. When the MR system **10** is in a receive mode, the switching unit **28** selectively couples the RF receiver **26** to the RF coil assembly **18**, **18'** and the ATMU **30** detects and compensates for impedance mismatches between the RF coil assembly **18**, **18'** and the RF receiver **26** or, in the multi-element RF coil assembly example, between each coil element of the RF coil assembly **18**, **18'** and each corresponding pre-amplifier of the RF receiver **26**.

With reference to FIGURE 3, each ATMU **30** includes a reflected power sensor **32** which detects the impedance mismatch. The reflected power sensor **32** can detect an impedance mismatch and magnitude various ways, e.g. by determining the voltage standing wave ratio (VSWR) or the like. The ATMU includes an inductor matrix **34** in series between the RF coil assembly **18**, **18'** and the RF transmitter/receiver **24**, **26** and a capacitor matrix **36** in parallel between the RF coil assembly **18**, **18'** and the RF transmitter/receiver **24**, **26**. The inductor matrix **34** and capacitor matrix **36** form a LC matching network, and it should be appreciated that the inductor matrix in parallel and the capacitor matrix in series between the RF coil assembly **18**, **18'** and the RF transmitter/receiver **24**, **26** is also contemplated.

A matrix controller **38** controls the inductor matrix **34** and capacitor matrix **36** to compensate for the detected impedance mismatch. The inductor matrix **34** includes a plurality of inductors of differing magnitudes, each inductor being coupled to a corresponding switch which selectively switches the inductor to one of an active or inactive state. Analogously, the capacitor matrix **36** includes a plurality of capacitors of differing magnitudes, each capacitor being coupled to a corresponding switch which selectively switches the inductor to one of an active or inactive state. The matrix controller **38** controls the switches of both the inductor and capacitor matrices **34**, **36**. In this manner, the inductor matrix **34** can elicit a plurality of inductances and the capacitor matrix **36** can elicit a plurality of capacitances. The matrix controller **38** also controls a network switch **40** which changes the configuration of the LC matching circuit to one of a rear L-network (FIGURE 4A), forward L-network (FIGURE 4B), or π-network (FIGURE 4C).

The matrix controller **38** includes a look-up table (LUT) **42** and a memory unit **44**. The look-up table associates a known impedance mismatch, which is stored on the memory unit **44**, to a corresponding inductor matrix program and/or capacitor matrix program, which are also stored on the memory unit **44**. It should be appreciated that a plurality of known impedances mismatches and corresponding inductor and/or capacitor matrix programs are stored on the memory unit **44**. The inductor and capacitor matrix programs instruct the matrix controller **38** which switches to control of the inductor and capacitor matrices **34**, **36** such that they compensate for the known impedance mismatch.

The matrix controller **38** compares the detected impedance mismatch to the known impedance mismatches stored on the memory unit **44**. If the detected impedance mismatch correlates to a known impedance mismatch, then the matrix controller programs the inductor matrix **34** and capacitor matrix **36** according to the inductor matrix program and capacitor matrix program, respectively, which correspond to the known impedance matrix.

If the detected impedance mismatch does not correlate to a stored known impedance mismatch, then the matrix controller **38** is configured to generate corresponding inductor and capacitor matrix programs by iterating through the plurality of inductances and capacitances elicited by the inductor and capacitor matrices **34**, **36** until the detected impedance mismatch is compensated for. The matrix controller **38** stores the detected impedance mismatch as a known impedance mismatch and the generated corresponding inductor and capacitor matrix programs and then updates the LUT **42** to include the association between the recently stored known impedance mismatch and corresponding inductor and capacitor matrix programs. In one embodiment, the matrix controller **38** determines which known impedance mismatch most closely correlates to the detected impedance mismatch or order to reduce the number of iterations required to compensate for the detected impedance mismatch. Using the closest correlation, the matrix controller **38** then systematically adjusts the corresponding inductor and capacitor matrix programs until the detected impedance mismatch is compensated. The detected impedance mismatch is stored as a known impedance mismatch onto memory unit **44** along with the corresponding inductor and capacitor matrix programs and the LUT **42** is updated. In this manner, the matrix controller configures a matching circuit from the available inductive and capacitive elements to provide a network that best matches the impedance of the coil **18**, **18’**.

The ATMU **30** is constructed of non-magnetic materials, e.g. the switches of the inductor and capacitor matrices are MEMS-based, pin-diodes, or the like. This allows the ATMU to be position relatively close to the RF coil assembly **18**, **18’** or the individual coil elements in a multi-element system and in turn allows to RF transmitter and/or receiver **24**, **26** to be positioned relatively close to the MR system **10**. With reference to FIGURE 5A, in one embodiment, the ATMU **30** is positioned adjacent to a transverse electromagnetic (TEM) coil **50**. With reference to FIGURE 5B, in another embodiment, the ATMU **30** is positioned adjacent to a loop coil **52**.

With reference to FIGURES 6A and 6B, in another embodiment, the ATMU **30** is integrated into the RF coil assembly. As illustrated, the ATMU 30 is integrated into a TEM coil **54** which is disposed in a recess **55** of a split gradient coil **56.** Spilt gradient coils allow for a larger bore size by defining a gap or recess **55** between gradient coils in which the RF coil assembly is disposed. The arrangement accommodates larger subject and may reduce anxiety. As shown in FIGURE 6B, a top-down view of the TEM coil with an integrated ATMU **30**, the ATMU can be disposed onto the TEM coil printed circuit element (PCB) **58.** Transmission lines connecting the ATMU **30** to the RF transmitter **24** and/or RF receiver **26** can be routed though the MR system **10** housing in such a manner as not to interfere with main magnet or gradient coil operation.

With reference to FIGURE 7, in a third aspect, the RF coil assembly **18**, **18'** is a multi-frequency, multi-element RF coil assembly, more specifically an interleaved double-tuned RF coil assembly. The RF coil assembly **18**, **18'** includes adjacent coil elements **60**, **62** which are tuned to different frequencies for simultaneous excitation of multiple nuclear species, e.g. 13C and 1H. However, other species, such as 31P, 19F, or the like, are also contemplated along with non-interleaved configurations and triple, quadruple, etc. tuned assemblies.

With reference to FIGURE 8, in a fourth aspect, the RF coil assembly **18**, **18'** is a multi-frequency, multi-element RF coil assembly, more specifically an interleaved double-tuned RF coil assembly where one array, e.g. 13C **60**, is connected via hardware combiner **64** to the RF transmitter **24.** The arrangement is useful when one of the multiple nuclear species does not present a patient dependent load on the individual coil elements due to the lower resonance frequency of 13C. In the example illustrated in FIGURE 8, 13C coil elements **60** are not connected to an ATMU because at a certain field strength, such as 2T, the 13C coil elements **60** will not present a significant patient dependent impedance mismatch, i.e. the subject dependent loading of the individual coil elements is not significant at the lower resonance frequency of 13C. However, at the same field strength, the 1H coil elements **62** will be loaded by the subject and present an impedance mismatch which can be compensated with the corresponding ATMUs **30.** It should be noted that other species, such as 31P, 19F, or the like, are also contemplated along with non-interleaved configurations and triple, quadruple, etc. tuned assemblies.

With reference to FIGURE 9, the scan controller **20** controls the ATMU(s) **30** to compensate for impedance mismatches in between image acquisitions as illustrated in the timing diagram. In one embodiment, prior to acquisition of any MR image representations, the scanner controller **20** controls the ATMU(s) **30** to compensate for impedance mismatches while the subject is in the optimal position in the examination region **14** such that all unknown impedance mismatches can be accounted for prior to image acquisition. In another embodiment, the scanner controller **20** controls the ATMU(s) **30** to compensate for impedance mismatches for moving bed examinations. In moving bed examinations, image representations of the subject are acquired at a number of different bed positions in the examination region **14**. Prior to the examination, the scanner controller **20** is configured to control the ATMU(s) **30** to compensation for impedance mismatches at all of the bed position such that all unknown impedance mismatches can be accounted for prior to image acquisition. Alternatively, the impedance mismatch can be measured during an MR scan and the ATMU(s) are adjusted dynamically during the scan to maintain optimal impedance matching.

With returning reference to FIGURE 1, the received data from the receivers **26** is temporarily stored in a data buffer **70** and processed by a magnetic resonance data processor **72**. The magnetic resonance data processor can perform various functions as are known in the art, including image reconstruction (MRI), magnetic resonance spectroscopy (MRS), catheter or interventional instrument localization, and the like. Reconstructed magnetic resonance images, spectroscopy readouts, interventional instrument location information, and other processed MR data are stored in memory, such as a medical facility's patient archive. A graphic user interface or display device **74** includes a user input device which a clinician can use for controlling the scan controller **20** to select scanning sequences and protocols, display MR data, and the like.

The invention has been described with reference to the preferred embodiments. Modifications and alterations may occur to others upon reading and understanding the preceding detailed description. It is intended that the invention be constructed as including all such modifications and alterations insofar as they come within the scope of the appended claims or the equivalents thereof.

CLAIMS

Having thus described the preferred embodiments, the invention is now claimed to be:

1. An automatic tune and match device (30), comprising:

a reflected power sensor (32) which detects power reflected from a load (18,18'); and

an LC matching circuit in series with the load, the LC matching circuit being programmable to minimize the reflected power.

2. The automatic tune and match device (30) according to claim 1, further including:

an inductor matrix (34) in series with the load (18,18');

a capacitor matrix (36) in parallel with the inductor matrix to minimize the reflected power; and

a matrix controller (38) which configures at least one of the inductor matrix or capacitor matrix based on the detected reflected power.

3. The automatic tune and match device (30) according to either one of claims 1 and 2, further including:

a network switch (40) which switches the LC matching circuit to one of a forward L-network, rear L-network, and a π-network parallel configuration.

4. The automatic tune and match device (30) according to any one of claims 1-3, further including:

a transmitter amplifier (24) connected with the load to supply power thereto; and

wherein matrix controller adjusts the inductor matrix (34) and capacitor matrix (36 to the impedance of the load and the transmitter amplifier.

5. The automatic tune and match device (30) according to any one of claims 2-4, wherein at least one of:

the inductor matrix (34) includes a plurality inductors of differing magnitudes, each inductor being coupled to a corresponding switch which selectively switches the inductor to one of an active or inactive state; and

the capacitor matrix (36) includes a plurality capacitors of differing magnitudes, each capacitor being coupled to a corresponding switch which switches the inductor to one of an active or inactive state.

6. The automatic tune and match device (30) according to any one claims 2-5, wherein the matrix controller (38) includes:

a look-up table (42) of a plurality known impedance mismatches, each known impedance mismatch being associated with a corresponding inductor matrix settings and capacitor matrix settings;

a memory element (44) which stores the known impedance mismatches and corresponding inductor and capacitor matrix programs.

7. The automatic tune and match device (30) according to claims 6, wherein the matrix controller (38) is configured to iteratively determine new inductor and capacitor matrix settings and store the new inductor and capacitor matrix settings in the memory element (44).

8. A magnetic resonance system (10), comprising:

a magnet (12) which generates a static magnetic field in an examination region (14);

a radiofrequency (RF) coil assembly (18, 18′) configured to induce and manipulate magnetic resonance in a subject in the examination region and/or acquire magnetic resonance data from the examination region (14);

an RF transmitter (24) coupled to the RF coil assembly (18,18’), the RF transmitter causes the RF coil assembly to generate the magnetic resonance; and

an automatic tune and match device (30) according to any one of claims 1-7, the automatic tune and match device being disposed between the RF transmitter (24) and RF coil assembly (18,18’) to match the impedance of the RF coil assembly (18,18’) and the RF transmitter (24).

9. The magnetic resonance system (10) according to claim 8, further including:

an RF receiver (26) coupled to the RF coil assembly (18,18'), the RF receiver being configured receive the generated magnetic resonance signals;

a switching unit (28) to selectively couple the RF coil assembly (18,18') to one of the RF transmitter (24) and RF receiver (26), the ATMU (30) being disposed one of:

(1) between the RF coil assembly and the switching unit (28) to match an impedance of the RF coil assembly (18,18') and an impedance of the RF receiver (26); and

(2) between the RF transmitter (24) and the switching unit.

10. The magnetic resonance system (10) according to either one of claims 8 and 9, further including:

a scan controller (20) which controls the automatic tune and match device (30) to detect and minimize reflected power at the RF coil assembly (18,18').

11. The magnetic resonance system (10) according to claim 10, wherein the scan controller (20) which controls the automatic tune and match device (30) to detect and minimize reflected power at the RF coil assembly (18,18') between image acquisitions.

12. The magnetic resonance system (10) according to any one of claims 8-10, wherein the automatic tune and match device (30) is integrated into the RF coil assembly (18,18').

13. The magnetic resonance system (10) according to claim 12, wherein the automatic tune and match device (30) integrated into the RF coil assembly (18,18') is disposed within a recess (55) of a gradient magnetic field coil (56).

14. A method for impedance matching, comprising:

detecting reflected power from a load; and

programming an LC matching circuit to minimize the reflected power.

15. The method according to claim 14, wherein configuring the LC matching circuit includes:

configuring an inductor matrix (34) in series with the load; and

configuring a capacitor matrix (36) in parallel with the inductor matrix to minimize the reflected power.

16. The method according to either one of claims 14 and 15, wherein programming the LC matching circuit includes:

switching the LC matching circuit to one of a forward L-network, rear L-network, and a π-network parallel configuration.

17. The method according to any one of claims 14-16, wherein programming the matching circuit further includes at least one of:

switching inductors of different magnitude of the inductor matrix (34) to one of an active or inactive state; and

switching capacitors of different magnitude of the capacitor matrix (36) to one of an active or inactive state.

18. The method according to any one of claims 14-17, wherein the load includes an RF coil assembly (18,18') of a magnetic resonance (MR) scanner (10) and a transmitter is connected to the LC matching circuit and further including:

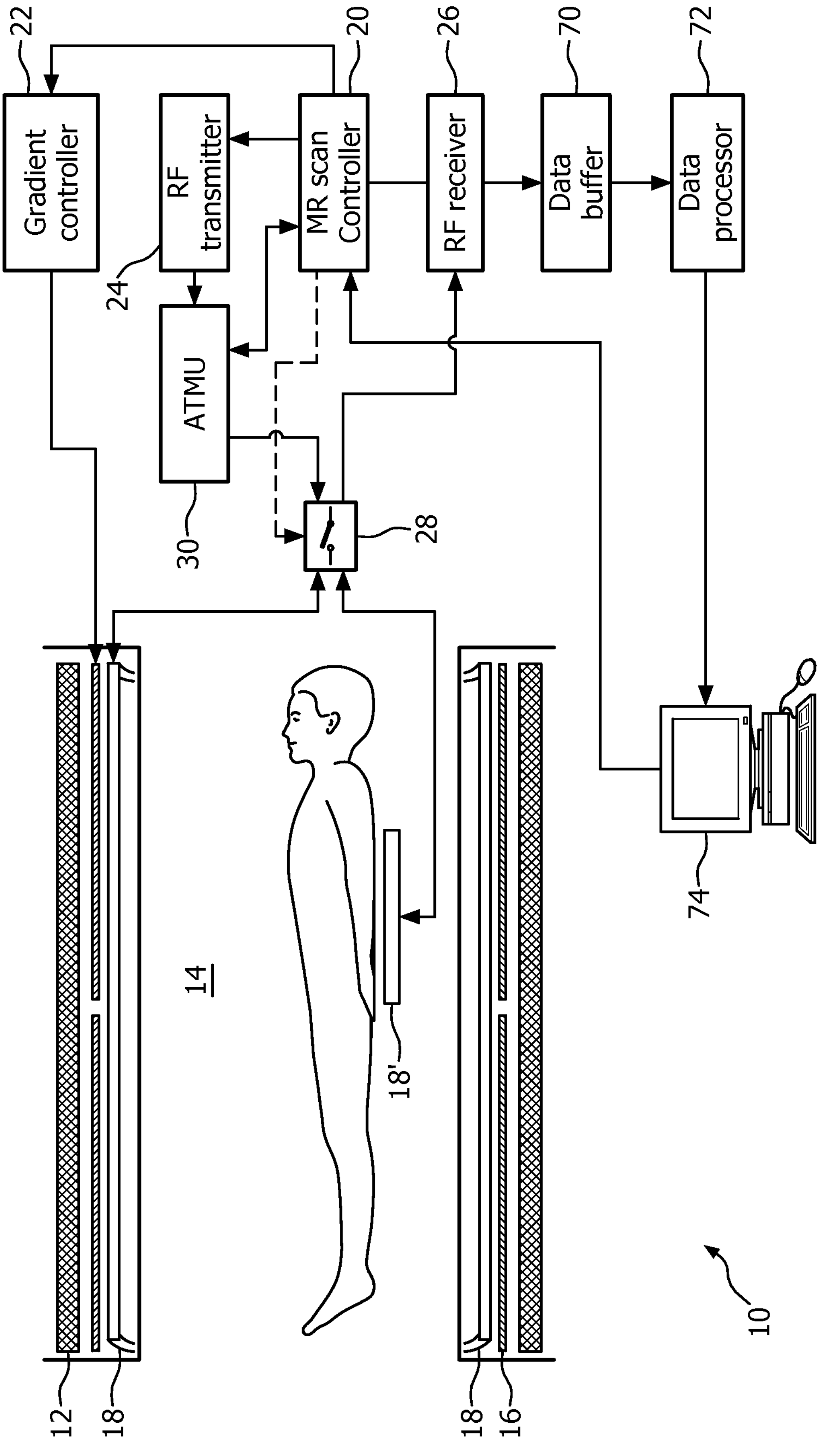
positioning a subject in an examination region (14) of the MR scanner adjacent the RF coil assembly; and

transmitting an RF pulse to the RF coil assembly via the LC matching circuit, the detecting step being performed on power reflected from the RF coil assembly in response to the RF pulse.

19. The method according to claim 18, further including:

conducting an MR examination sequence;

moving the subject relative to the RF coil assembly (18,18') during the MR examination sequence; and during the MR examination sequence, repeating the reflected power detecting step and the matching circuit programming step.


22
Gradient controller
RF transmitter
24
20
MR scan Controller
26
RF receiver
70
Data buffer
72
Data processor
ATMU
30
28
14
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74
12
18
18
16
10

FIG. 1

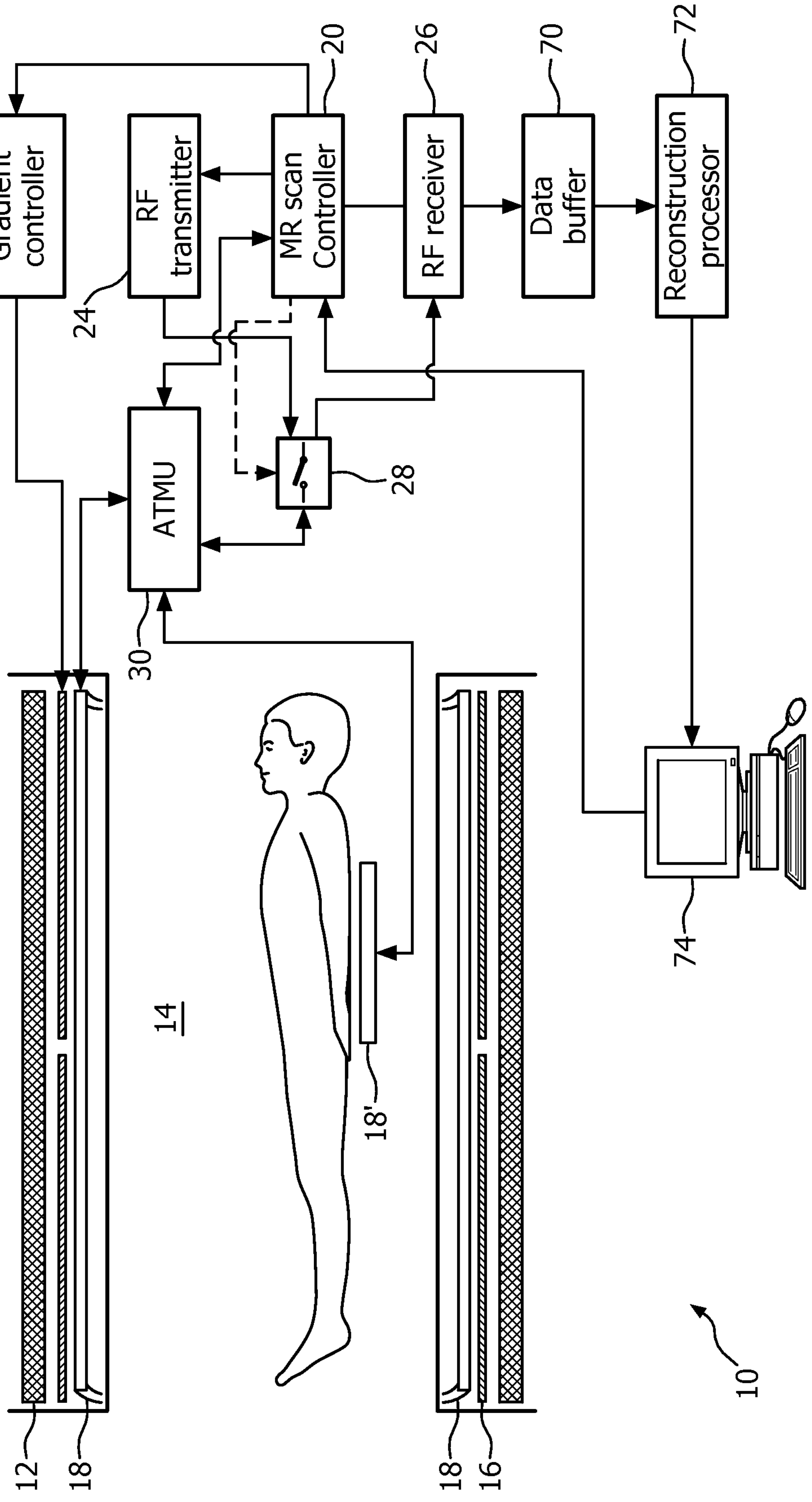
Gradient controller
RF transmitter
MR scan Controller
RF receiver
Data buffer
Reconstruction processor
ATMU
22
24
20
26
70
72
28
30
14
18'
74
10
12
18
18
16

FIG. 2

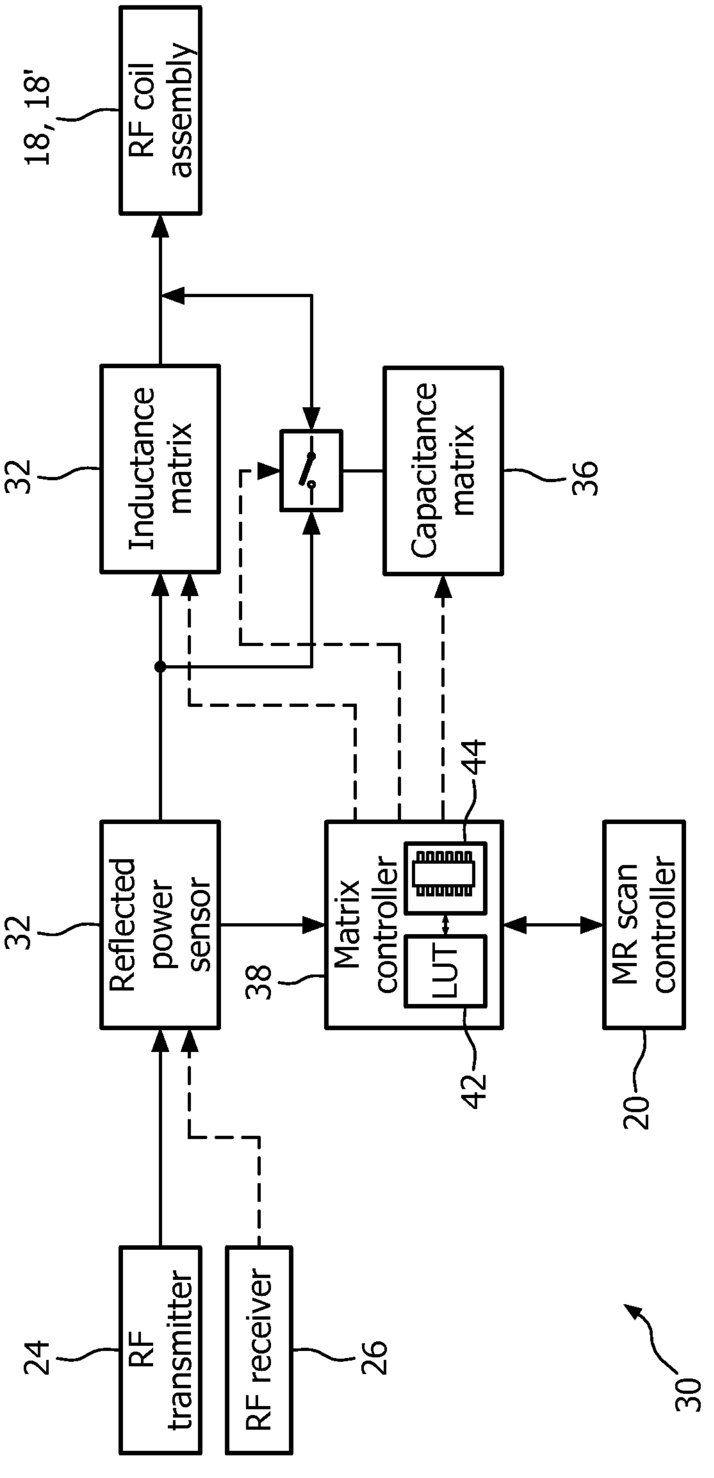
18, 18'
RF coil assembly
32
Inductance matrix
36
Capacitance matrix
32
Reflected power sensor
38
Matrix controller
LUT
42
44
MR scan controller
20
24
RF transmitter
RF receiver
26
30

FIG. 3

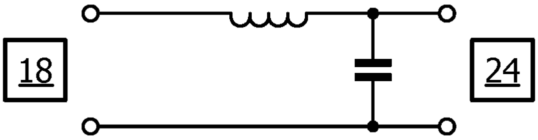

FIG. 4A

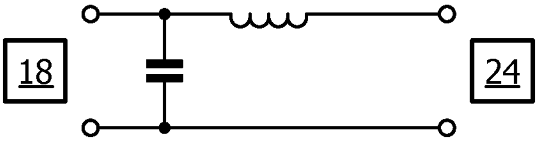

FIG. 4B

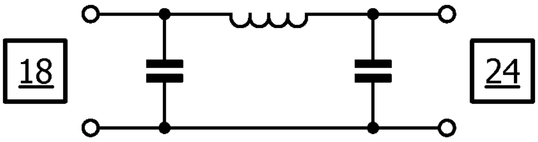

FIG. 4C

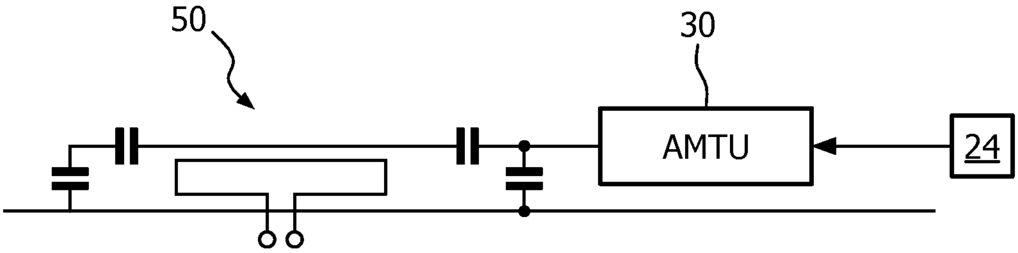

FIG. 5A

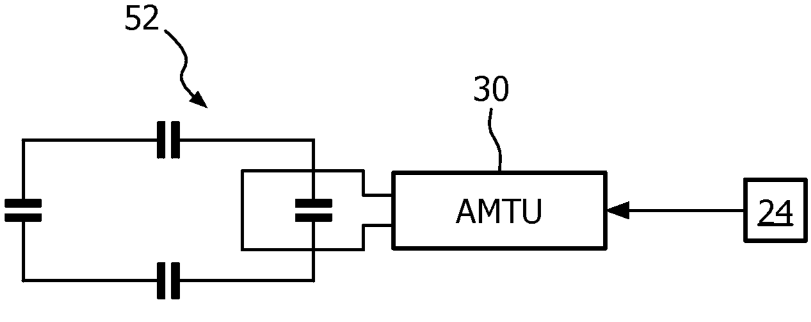

FIG. 5B

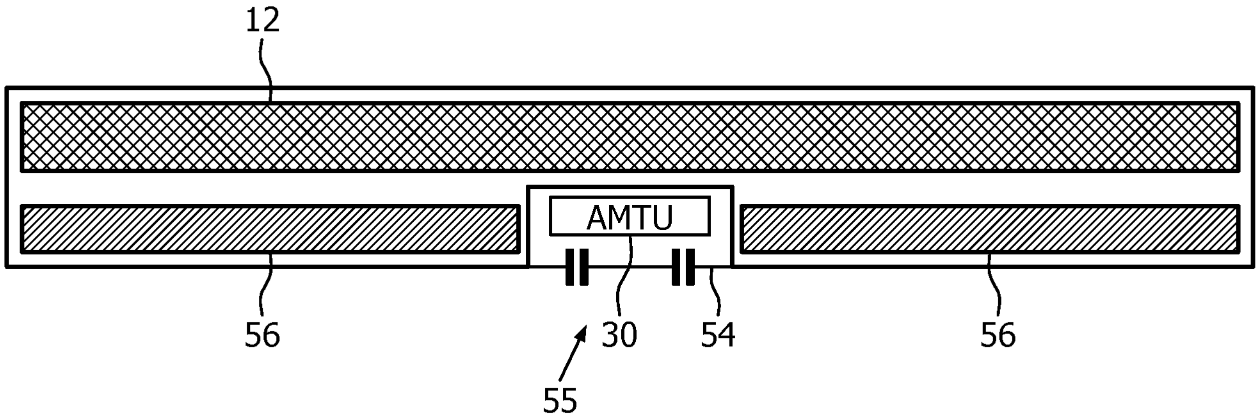

FIG. 6A

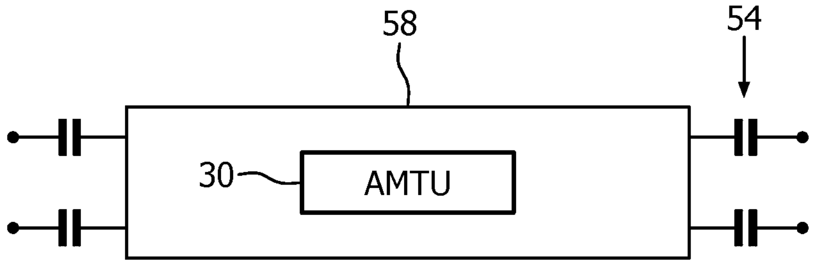

FIG. 6B

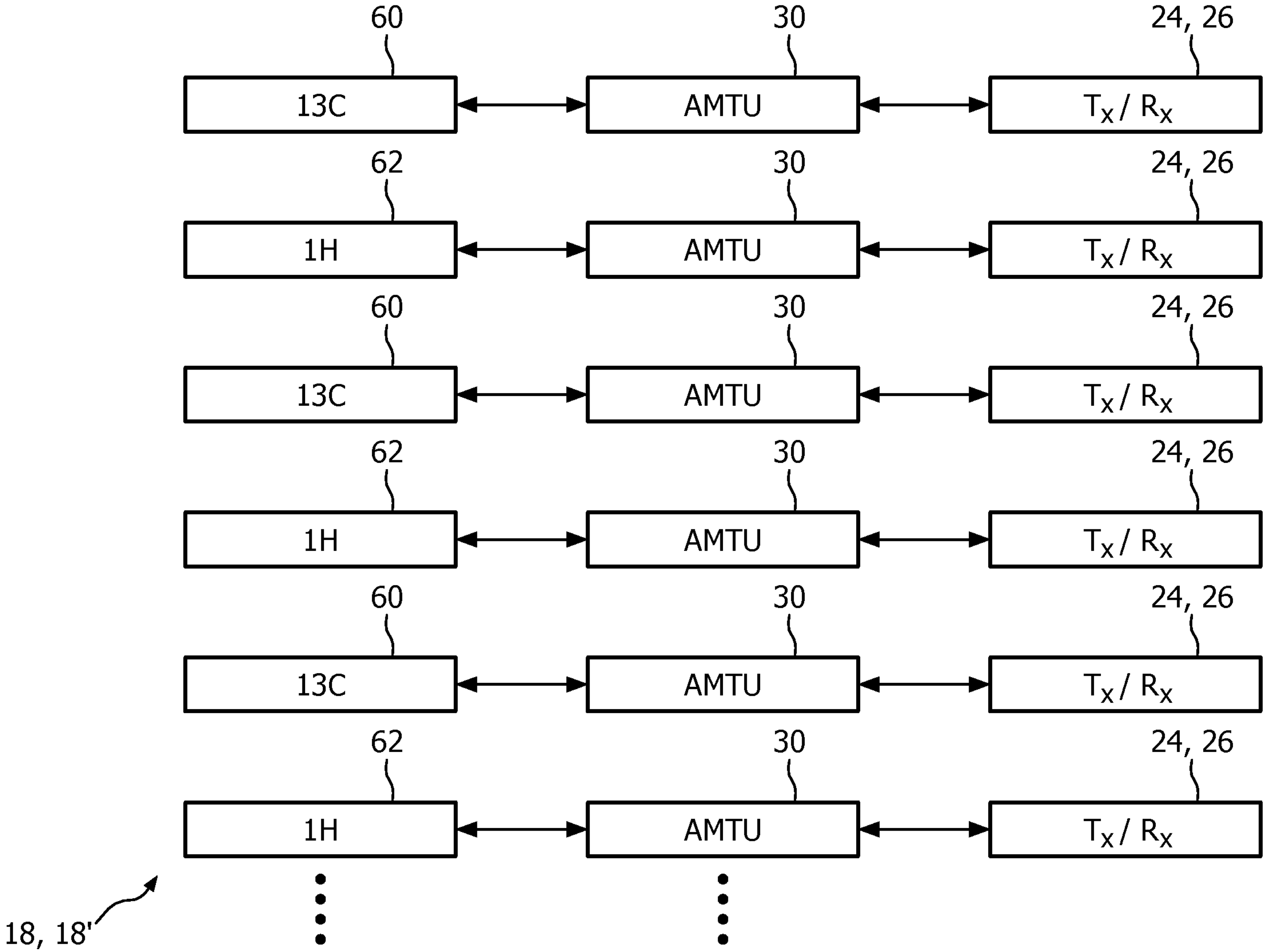
60
30
24, 26
13C
AMTU
T_x / R_x
62
30
24, 26
1H
AMTU
T_x / R_x
60
30
24, 26
13C
AMTU
T_x / R_x
62
30
24, 26
1H
AMTU
T_x / R_x
60
30
24, 26
13C
AMTU
T_x / R_x
62
30
24, 26
1H
AMTU
T_x / R_x
18, 18'

FIG. 7

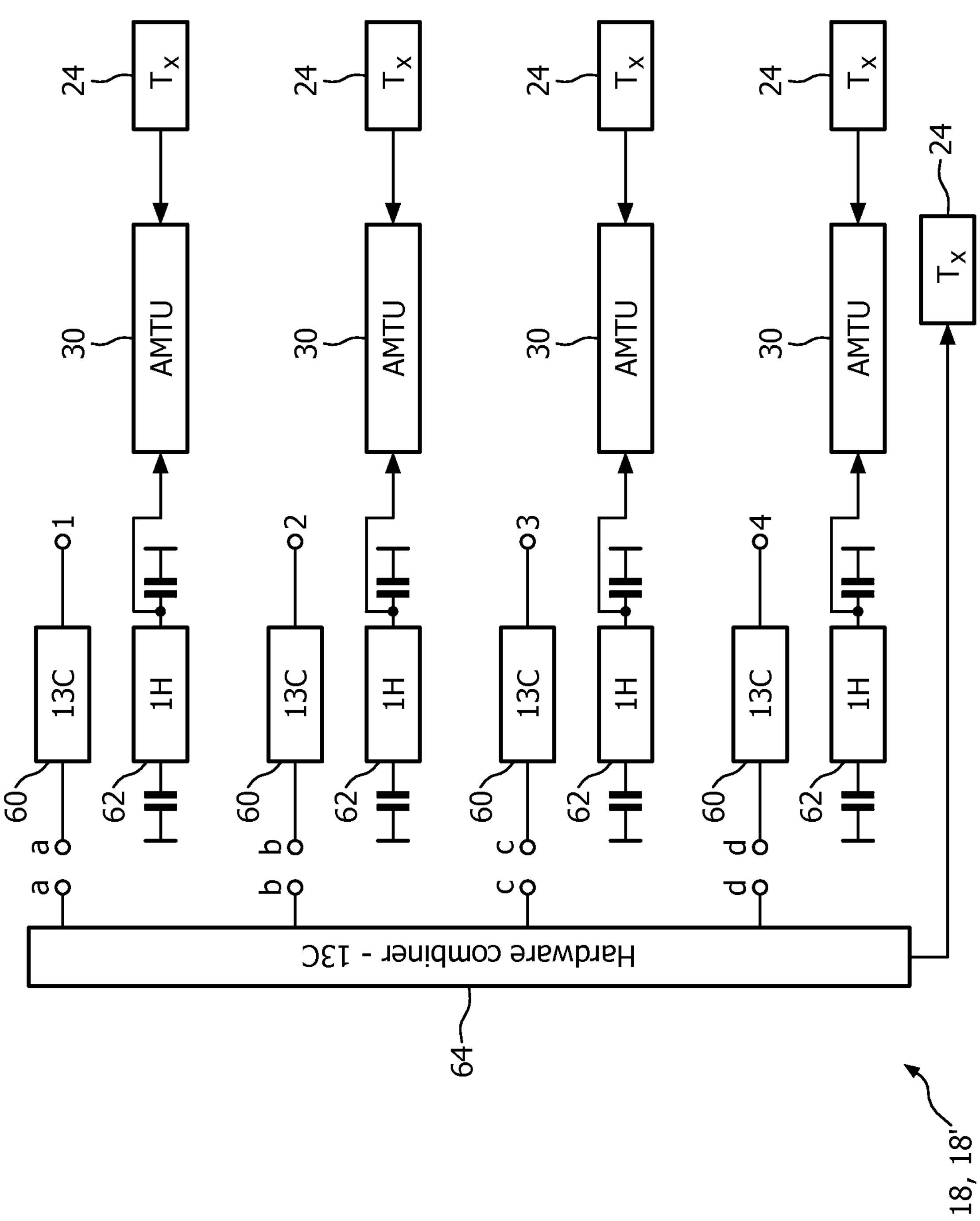
24
Tx
30
AMTU
24
Tx
30
AMTU
24
Tx
30
AMTU
24
Tx
30
AMTU
24
Tx
1
2
3
4
13C
1H
60
62
a
b
c
d
Hardware combiner - 13C
64
18, 18'

FIG. 8

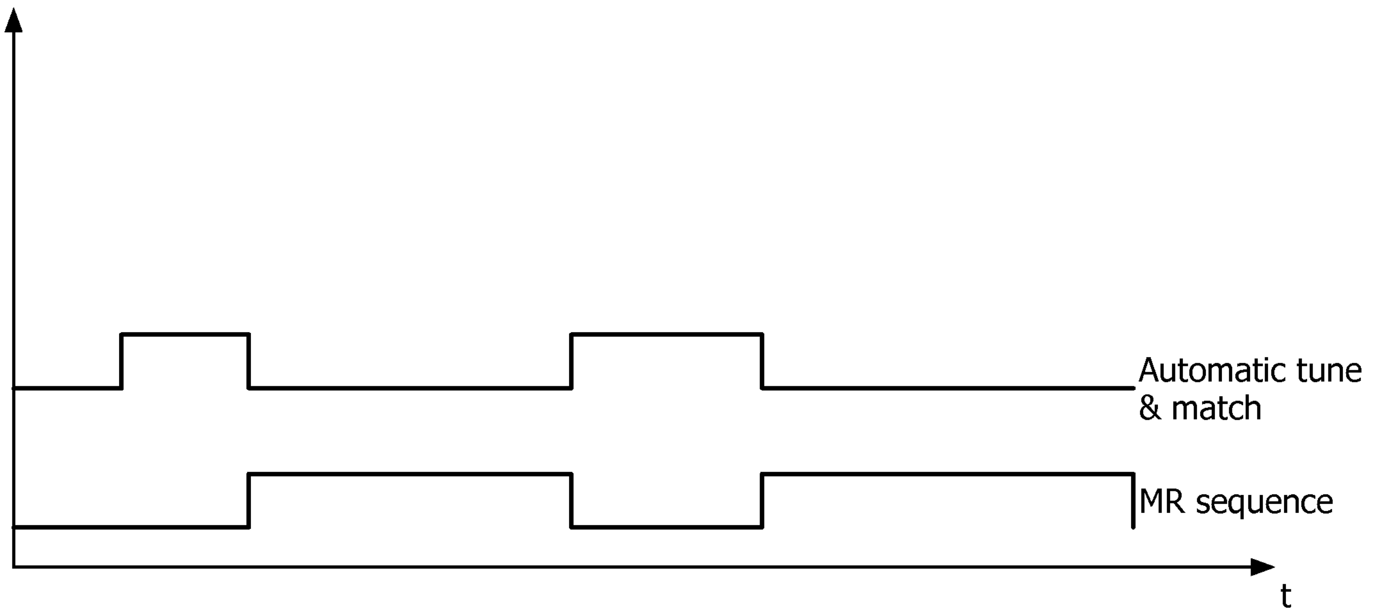

FIG. 9

INTERNATIONAL SEARCH REPORT

International application No
PCT/IB2010/053553

A. CLASSIFICATION OF SUBJECT MATTER
INV. G01R33/36
ADD.

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

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
G01R

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

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

EPO-Internal, INSPEC

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	US 5 300 068 A (ROSAR GEORGE C [US] ET AL) 5 April 1994 (1994-04-05) column 6, line 24 - column 8, line 63; figures 1,3,4	1-7, 14-17
X	WO 88/08645 A1 (BR COMMUNICATIONS [US]) 3 November 1988 (1988-11-03) page 1, line 1 - page 7, line 8; figure 1	1,2,4-7, 14,15,17

-/--

☒ Further documents are listed in the continuation of Box C.

☒ See patent family annex.

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Date of the actual completion of the international search 25 October 2010	Date of mailing of the international search report 29/10/2010
Name and mailing address of the ISA/ European Patent Office, P.B. 5818 Patentlaan 2 NL - 2280 HV Rijswijk Tel. (+31-70) 340-2040, Fax: (+31-70) 340-3016	Authorized officer Lersch, Wilhelm

C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
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Y	column 1, line 10 - column 2, line 10 column 2, line 50 - column 5, line 60 figures 1-5 -----	13
X	US 6 252 461 B1 (RAAB FREDERICK HERBERT [US]) 26 June 2001 (2001-06-26) column 6, line 59 - column 7, line 2 column 7, line 35 - column 7, line 52 column 13, line 7 - column 13, line 34 column 16, line 11 - column 16, line 26 column 21, line 6 - column 21, line 16 figures 10,14,20,21 -----	1-12, 14-19
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Y	US 2005/099183 A1 (HEID OLIVER [DE] ET AL HEID OLIVER [DE] ET AL) 12 May 2005 (2005-05-12) paragraphs [0006] - [0008], [0021] - [0039] figures 1-4 ----- -/--	13

C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
Y	KURPAD K N ET AL: "RF current element design for independent control of current amplitude and phase in transmit phased arrays" CONCEPTS IN MAGNETIC RESONANCE, NMR CONCEPTS, KINGSTON, RI, US, vol. 29B, 5 April 2006 (2006-04-05), pages 75-83, XP002457869 ISSN: 1043-7347	13
A	see the chapter 'Design and Construction' -----	12
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International application No
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