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(54) **Intracavity probes and interfaces therefor for use in obtaining images and spectra of intracavity structures using high field magnetic resonance systems**

Intrakavitätssonden und Schnittstellen dafür zur Verwendung beim Erhalt von Bildern und Spektren von Interkavitätsstrukturen unter Verwendung von Hochfeld-Magentresonanzsystemen

Sondes intracavitaires et interfaces associées utilisées pour obtenir des images et des spectres de structures intracavitaires au moyen de systèmes de résonance magnétique de haute résolution

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Description**FIELD OF THE INVENTION**

5 **[0001]** The present invention generally relates to systems for obtaining images and spectra of intracavity structures using magnetic resonance (MR) systems. More particularly, the invention pertains to an intracavity probe capable of being inserted into various bodily openings, such as the rectum, vagina, mouth, etc., to obtain high resolution images of and spectroscopic results for regions of interest therein. Even more particularly, the invention relates to interface devices designed to interface such an intracavity probe with 2.0 Tesla to 5.0 Tesla MR systems to obtain such high
10 resolution images and spectroscopic results for such regions of interest.

BRIEF DESCRIPTION OF RELATED ART

15 **[0002]** The following background information is provided to assist the reader to understand the invention disclosed below and the environment in which it will typically be used. The terms used herein are not intended to be limited to any particular narrow interpretation unless clearly stated otherwise, either expressly or impliedly, in this document.

[0003] Magnetic resonance imaging (MRI) is a noninvasive method of producing high quality images of the interior of the human body. It allows medical personnel to see inside the human body without surgery or the use of ionizing radiation such as X-rays. The images are of such high resolution that disease and other forms of pathology can often be visually
20 distinguished from healthy human tissue. Magnetic resonance techniques and systems have also been developed for performing spectroscopic analyses by which the chemical content of tissue or other material can be ascertained.

[0004] MRI uses a powerful magnet, radio waves and computer technology to create detailed images of the soft tissues, muscles, nerves and bones in the human body. It does so by taking advantage of a basic property of the hydrogen atom, an atom that is found in abundance in all cells within the human body. In the absence of a magnetic field, the
25 nuclei of hydrogen atoms spin like a top, or precess, randomly in every direction. When subject to a strong magnetic field, however, the spin-axes of the hydrogen nuclei align themselves in the direction of that field. This is because the nucleus of the hydrogen atom has what is referred to as a large magnetic moment, which is basically a strong inherent tendency to line up with the direction of the magnetic field. Collectively, the hydrogen nuclei of the area to be imaged create an average vector of magnetization that points parallel to the magnetic field.

30 **[0005]** A typical MRI system, or scanner, includes a main magnet, three gradient coils, a radio frequency (RF) antenna (often referred to as the whole body coil), and a computer station from which an operator can control the overall MRI system. The chief component of the MRI system, however, is the main magnet. It is typically superconducting in nature and cylindrical in shape. Within its cylindrical bore (into which patients are placed during an MRI procedure), the main magnet generates a strong magnetic field, often referred to as the B_0 field, which is both uniform and static (non-varying).
35 This B_0 magnetic field is oriented along the longitudinal axis of the bore, referred to as the z direction, which compels the magnetization vectors of the hydrogen nuclei in the body to align themselves in that direction. In this alignment, the nuclei are prepared to receive RF energy of the appropriate frequency from the whole body coil. This frequency is known as the Larmor frequency and is governed by the equation $\omega = \gamma B_0$, where ω is the Larmor frequency (at which the hydrogen atoms precess), γ is the gyromagnetic constant, and B_0 is the strength of the magnetic field.

40 **[0006]** The RF antenna, or whole body coil, is generally used both to transmit pulses of RF energy and to receive the resulting magnetic resonance (MR) signals induced thereby in the hydrogen nuclei. Specifically, during its transmit cycle, the body coil broadcasts RF energy into the cylindrical bore. This RF energy creates a radio frequency magnetic field, also known as the RF B_1 field, whose magnetic field lines are directed in a line perpendicular to the magnetization vector of the hydrogen nuclei. The RF pulse (or B_1 field) causes the spin-axes of the hydrogen nuclei to tilt with respect to the
45 main (B_0) magnetic field, thus causing the net magnetization vector to deviate from the z direction by a certain angle. The RF pulse, however, will affect only those hydrogen nuclei that are precessing about their axes at the frequency of the RF pulse. In other words, only the nuclei that "resonate" at that frequency will be affected, and such resonance is achieved in conjunction with the operation of the three gradient coils.

50 **[0007]** The gradient coils are electromagnetic coils. Each gradient coil is used to generate a linearly varying yet static magnetic field along one of the three spatial directions (x,y,z) within the cylindrical bore known as the gradient H_1 field. Positioned inside the main magnet, the gradient coils are able to alter the main magnetic field on a very local level when they are turned on and off very rapidly in a specific manner. Thus, in conjunction with the main magnet, the gradient coils can be operated according to various imaging techniques so that the hydrogen nuclei at any given point or in any given strip, slice or unit of volume-- will be able to achieve resonance when an RF pulse of the appropriate frequency
55 is applied. In response to the RF pulse, the precessing hydrogen atoms in the selected region absorb the RF energy being transmitted from the body coil, thus forcing the magnetization vectors thereof to tilt away from the direction of the main (B_0) magnetic field. When the body coil is turned off, the hydrogen nuclei begin to release the RF energy in the form of the MR signal, as explained further below.

[0008] One well known technique that can be used to obtain images is referred to as the spin echo imaging technique. Operating according to this technique, the MRI system first activates one gradient coil to set up a magnetic field gradient along the z-axis. This is called the "slice select gradient," and it is set up when the RF pulse is applied and is shut off when the RF pulse is turned off. It allows resonance to occur only within those hydrogen nuclei located within a slice of the area being imaged. No resonance will occur in any tissue located on either side of the plane of interest. Immediately after the RF pulse ceases, all of the nuclei in the activated slice are "in phase," i.e., their magnetization vectors all point in the same direction. Left to their own devices, the net magnetization vectors of all the hydrogen nuclei in the slice would relax, thus realigning with the z direction. Instead, however, the second gradient coil is briefly activated to create a magnetic field gradient along the y-axis. This is called the "phase encoding gradient." It causes the magnetization vectors of the nuclei within the slice to point, as one moves between the weakest and strongest ends of the gradient, in increasingly different directions. Next, after the RF pulse, slice select gradient and phase encoding gradient have been turned off, the third gradient coil is briefly activated to create a gradient along the x-axis. This is called the "frequency encoding gradient" or "read out gradient," as it is only applied when the MR signal is ultimately measured. It causes the relaxing magnetization vectors to be differentially re-excited, so that the nuclei near the low end of the gradient begin to precess at a faster rate, and those at the high end pick up even more speed. When these nuclei relax again, the fastest ones (those which were at the high end of the gradient) will emit the highest frequency of radio waves.

[0009] Collectively, the gradient coils allow the MR signal to be spatially encoded, so that each portion of the area being imaged is uniquely defined by the frequency and phase of its resonance signal. In particular, as the hydrogen nuclei relax, each becomes a miniature radio transmitter, giving out a characteristic pulse that changes over time, depending on the local microenvironment in which it resides. For example, hydrogen nuclei in fats have a different microenvironment than do those in water, and thus transmit different pulses. Due to these differences, in conjunction with the different water-to-fat ratios of different tissues, different tissues transmit radio signals of different frequencies. During its receive cycle, the body coil detects these miniature radio transmissions, which are often collectively referred to as the MR signal. From the body coil, these unique resonance signals are conveyed to the receivers of the MR system where they are converted into mathematical data corresponding thereto. The entire procedure must be repeated multiple times to form an image with a good signal-to-noise ratio (SNR). Using multidimensional Fourier transformations, an MR system can convert the mathematical data into a two- or even a three-dimensional image.

[0010] When more detailed images of a specific part of the body are needed, a local coil is often used in addition to, or instead of, the whole body coil. A local coil can take the form of a volume coil or a surface coil. A volume coil is used to surround or enclose the volume to be imaged (e.g., a head, an arm, a wrist, a leg, a knee or other region of interest). A surface coil, however, is merely fitted or otherwise placed against a particular surface of the patient so that the underlying region of interest can be imaged (e.g., the abdominal, thoracic and/or pelvic regions). In addition, a local coil can be designed to operate either as a receive-only coil or a transmit/receive (T/R) coil. A receive-only coil is only capable of detecting the MR signals produced by the human body (in response to the B_1 magnetic field generated by the MR system during a scanning procedure). A T/R coil, however, is capable of both receiving the MR signals as well as transmitting the RF pulses that produce the RF B_1 magnetic field, which is the prerequisite for inducing resonance in the tissues of the region of interest.

[0011] It is well known in the field of MRI to use a single local coil, whether surface or volume, to detect the MR signals. According to the single coil approach, a relatively large local coil is used to cover or enclose the entire region of interest. Early receiving coils were just linear coils, meaning that they could detect only one of the two (i.e., vertical M_x and horizontal M_y) quadrature components of the MR signals produced by the region of interest. Subsequent receiving coils, however, employed quadrature mode detection, meaning that they could intercept both the vertical and horizontal components. Compared to linear receiving coils, quadrature receiving coils enabled MRI systems to provide images for which the SNR was much improved, typically by as much as 41%. Even with the improvement brought with quadrature mode detection, the single coil approach still provided images whose quality invited improvement. The disadvantage inherent to the single coil approach is attributable to just one coil structure being used to acquire the MR signals over the entire region of interest.

[0012] Phased array coils were developed to overcome the shortcomings with the single coil approach. Instead of one large local coil, the phased array approach uses a plurality of smaller local coils, with each such coil covering or enclosing only a portion of the region of interest. In a system having two such coils, for example, each of the coils would cover or enclose approximately half of the region of interest, with the two coils typically being partially overlapped for purposes of magnetic isolation. The two coils would acquire the MR signals from their respective portions simultaneously, and they would not interact adversely due to the overlap. Because each coil covers only half of the region of interest, each such coil is able to receive the MR signals at a higher SNR ratio for that portion of the region of the interest within its coverage area. The smaller local coils of the phased array thus collectively provide the MRI system with the signal data necessary to generate an image of the entire region of interest that is higher in resolution than what can be obtained from a single large local coil.

[0013] One example of a phased array coil is the Gore[®] torso array produced by W.L. Gore and Associates, Inc. The

torso array contains four surface coils, two of which disposed in an anterior paddle and the other two disposed in a posterior paddle. The two paddles are designed to be placed against the anterior and posterior surfaces, respectively, of the patient about the abdominal, thoracic and pelvic regions. The torso array is designed for use with an MR system whose data acquisition system has multiple receivers. The four leads of the torso array, one each from the two anterior surface coils and the two posterior surface coils, can be connected to separate receivers, with each receiver amplifying and digitizing the signal it receives. The MR system then combines the digitized data from the separate receivers to form an image whose overall SNR is better than what could be obtained from a single local coil, or even two larger anterior and posterior local coils, covering the entire region of interest alone.

[0014] It is also well known to obtain images of internal bodily structures through the use of intracavity probes. An example of a prior art intracavity probe can be found in U.S. Patents 5,476,095 and 5,355,087, both of which are assigned to the assignee of the present invention and incorporated herein by reference. The prior art probe disclosed in those patents is designed to be inserted into bodily openings such as the rectum, vagina, and mouth. Those patents also disclose interface devices that are designed to interface the prior art intracavity probe with MR imaging and spectroscopy systems. A method of using the intracavity probe is disclosed in U.S. Patent 5,348,010, which is also assigned to the assignee of the present invention and incorporated herein by reference.

[0015] The prior art probe, operated in conjunction with its associated interface unit, allows an MR system to generate images of, and spectroscopic results for, various internal bodily structures such as the prostate gland, colon or cervix. Examples of such prior art probes include the BPX-15 prostate/endorectal coil (E-coil), the PCC-15 colorectal coil, and the BCR-15 cervix coil, all of which are part of the MRInnervu[®] line of disposable coils produced by MEDRAD, Inc. of Indianola, Pennsylvania. Examples of such interface units include the ATD-II and the ATD-Torso units, also produced by MEDRAD, Inc.

[0016] The ATD-II unit is used to interface the prior art probe with one receiver of an MR system to provide images or spectra of the region of interest, namely, the prostate gland, colon or cervix. The ATD-Torso unit is used to interface not only the prior art probe but also the Gore[®] torso array with multiple receivers of the MR system. When connected to such a probe and the torso array, the ATD-Torso unit allows the MR system to provide images or spectra not only of the prostate gland, colon or cervix but also of the surrounding anatomy, i.e., the abdominal, thoracic and pelvic regions.

[0017] Despite their widespread acceptance and good reputation in the marketplace, these prior art intracavity probes and interfaces units nevertheless have a few shortcomings. First, the prior art probe and its associated interface units (i.e., ATD-II and ATD Torso units) are designed to operate only with 1.0 or 1.5 Tesla MR systems. Consequently, they are not suitable for use with MR systems designed to operate at higher field strengths, such as the 2.0 to 5.0 Tesla and particularly 3.0 Tesla MR systems that are capable of producing even higher quality images and spectrographic results. Second, as a result of that design constraint, the prior art intracavity probe was designed with a coil loop that exhibits a 750 to 1000 ohm output impedance. Consequently, the interface units for the prior art probe had to include a π network or similar circuitry to match the high output impedance of the coil loop to the low input impedance (e.g., 50 ohms) required by various MR systems. Third, the design of the prior art probe allowed the tuning of its coil loop to deviate from the operating frequency of the MR system, the extent to which depending on the particular conditions (e.g., patients) in which the probe was used. Therefore, the prior art interface units for the prior art probe typically had to include tuning circuitry so as to assure that the intracavity probe could be tuned to the operating frequency of the MR system under all operating conditions.

[0018] WO 03/098236 A discloses an MR System featuring an intracavity probe and an interface device associated therewith.

[0019] US 4 928 064 A discloses a surface coil for use in MRI data acquisition systems.

SUMMARY OF THE INVENTION

[0020] Several objectives of the invention and other advantages are attained by an interface device for interfacing an intracavity probe with a magnetic resonance system having the features of claim 1.

[0021] In one aspect of a presently preferred embodiment, the invention provides an intracavity probe for use with a magnetic resonance (MR) system for obtaining images or spectra of a region of interest within a cavity of a patient. The intracavity probe includes a first coil loop and first and second output cables. Designed to receive MR signals from the region of interest, the first coil loop has a plurality of capacitors including first and second drive capacitors and a tuning capacitor. The first and second drive capacitors are serially connected within the first coil loop and at a junction node thereof form a virtual ground for electrically balancing and impedance matching the first coil loop. The first and second drive capacitors are of approximately equal value. The tuning capacitor is serially connected within the first coil loop diametrically opposite the junction node of the two drive capacitors. The tuning capacitor has a value selected to resonate the first coil loop at an operating frequency of the MR system. The first output cable is connected at one end thereof across the first drive capacitor, and the second output cable is connected at one end thereof across the second drive capacitor. The first and second output cables each have an electrical length of $S_L + n(\lambda/4)$ wherein S_L is a supplemental

length whose reactance is of a same magnitude as a reactance of the drive capacitor corresponding thereto, n is an odd integer, and λ is a wavelength of the operating frequency. The two output cables terminate in a plug therefor for connecting the first coil loop to an interface device for the intracavity probe.

5 [0022] In another aspect of the presently preferred embodiment, the invention provides an interface device for interfacing an intracavity probe with a (probe) input port of a magnetic resonance (MR) system which is not equipped with its own preamplifier. The MR system has a receive cycle and a transmit cycle of operation, and the intracavity probe has a coil loop and a pair of output cables for connecting the coil loop to the interface device. The interface device includes a phase shifting network and a preamplifier. During the receive cycle, the phase shifting network enables the coil loop to be coupled through the output cables to the probe input port of the MR system and permits MR signals received from each of the output cables to be constructively combined. During the transmit cycle, the phase shifting network enables the coil loop to be decoupled through the output cables from a transmit field of the MR system. The preamplifier provides gain and impedance matching between the phase shifting network and the probe input port so that the MR signals received from the phase shifting network are passed with enhancement of signal-to-noise ratio to the probe input port during the receive cycle.

15 [0023] In a related aspect of the presently preferred embodiment, the invention provides an interface device for interfacing an intracavity probe with a (probe) input port of a magnetic resonance (MR) system which is equipped with its own preamplifier. The MR system has a receive cycle and a transmit cycle of operation, and the intracavity probe has a coil loop and a pair of output cables for connecting the coil loop to the interface device. The interface device includes a phase shifting network. During the receive cycle, the phase shifting network enables the coil loop to be coupled through the output cables to the probe input port of the MR system and permits MR signals received from each of the output cables to be constructively combined and routed to the probe input port. During the transmit cycle, the phase shifting network enables the coil loop to be decoupled through the output cables from a transmit field of the MR system.

20 [0024] In yet another aspect of the presently preferred embodiment, the invention provides an interface device for interfacing both an intracavity probe and a coil system with a magnetic resonance (MR) system. The MR system has a receive cycle and a transmit cycle of operation, and the intracavity probe has a coil loop and a pair of output cables for connecting the coil loop to the interface device. The interface device includes a phase shifting network and an array interface circuit. During the receive cycle, the phase shifting network enables the coil loop to be coupled through the output cables to a probe input port of the MR system and permits MR signals received from each of the output cables to be constructively combined and routed to the probe input port. During the transmit cycle, the phase shifting network enables the coil loop to be decoupled through the output cables from a transmit field of the MR system. The array interface circuit is for electrically interconnecting one or more coil elements of the coil system and other input ports of the MR system.

25 [0025] In a variation of the presently preferred embodiment, the disclosure provides an intracavity probe for use with a magnetic resonance (MR) system for obtaining images or spectra of a region of interest within a cavity of a patient. The intracavity probe includes a first coil loop and an output cable. Designed to receive MR signals from the region of interest, the first coil loop has a plurality of capacitors serially connected therein. The plurality of capacitors includes a drive capacitor and a tuning capacitor. The drive capacitor is used for electrically balancing and impedance matching the first coil loop. The tuning capacitor is positioned diametrically opposite the drive capacitor and has a value selected to resonate the first coil loop at an operating frequency of the MR system. An output cable is connected at one end thereof across the drive capacitor. The output cable has an electrical length of $S_L + n(\lambda/4)$ wherein S_L is a supplemental length whose reactance is of a same magnitude as a reactance of the drive capacitor, n is an odd integer, and λ is a wavelength of the operating frequency. The output cable terminates in a plug therefor for connecting the first coil loop to an interface device for the intracavity probe.

30 [0026] It should be understood that the present invention is not limited to the presently preferred embodiment(s) and related aspects discussed above.

BRIEF DESCRIPTION OF THE DRAWINGS

35 [0027] The invention and its presently preferred and alternative embodiments will be better understood by reference to the detailed disclosure below and to the accompanying drawings, wherein:

40 **Figure 1** is a schematic diagram of a coil loop and an output cable of an intracavity probe according to one aspect of a first embodiment of the invention;

45 **Figure 2** is a perspective view showing the intracavity probe of Figure 1 in its fully assembled and fully equipped state;

Figure 3 is a cross-sectional view of the intracavity probe taken through line 3-3 of Figure 2 showing a distal end of the probe and inflatable balloon(s) attached thereto;

Figure 4 is partial cross-sectional view of the intracavity probe taken through line 4-4 of Figure 2 showing its shaft in cross-section and the two lumens defined therein and an anti-migration disc snapped onto the shaft;

5 **Figure 5** is a cross-sectional view of the distal end of the intracavity probe taken through line 5-5 of Figure 3 showing its outer and inner balloons, its coil loop situated between the balloons, and its shaft with the two lumens defined therein;

10 **Figure 6** is a cross-sectional view of the distal end of the intracavity probe taken through line 6-6 of Figure 3 showing its coil loop situated atop an anterior surface of the inner balloon;

Figure 7 is a cross-sectional view of the shaft of the intracavity probe of Figure 2 illustrating the two lumens defined therein and a flexible tip at its distal end;

15 **Figure 8** is a schematic diagram of an interface device according to another aspect of the first embodiment of the invention wherein, in its single-receiver version, the interface device has a probe interface circuit for interfacing the intracavity probe of

20 Figures 1-7 with a (probe) input port of a magnetic resonance (MR) system which is not equipped with its own preamplifier;

Figure 9 is a schematic diagram of an interface device according to yet another aspect of the first embodiment of the invention wherein, in its multiple-receiver version, the interface device has (i) a probe interface circuit for interfacing the intracavity probe of

25 Figures 1-7 with a (probe) input port of an MR system equipped with its own preamplifier and (ii) an array interface circuit for interfacing a phased array coil system, such as the Gore[®] torso array, with the (coil) input ports of the MR system;

30 **Figure 10** is a perspective view of the interface device in its single-receiver version of Figure 8, which is designed to interface the intracavity probe to the MR system via a (probe) input port thereof that is not equipped with a preamplifier;

35 **Figure 11** is a perspective view of the interface device in its multiple-receiver version of Figure 9, which is designed to interface the intracavity probe and a phased array coil system, such as the Gore[®] torso array, with the Phased Array Port of the MR system;

Figure 12 is a schematic diagram of a coil loop and an output cable of an intracavity probe, and a decoupling diode of an interface device corresponding thereto, according to a first alternative embodiment of the invention;

40 **Figure 13** is a schematic diagram of a coil loop and an output cable of an intracavity probe, and decoupling diodes of an interface device corresponding thereto, according to a second alternative embodiment of the invention;

Figure 14 is a schematic diagram of a coil loop and an output cable of an intracavity probe, and a decoupling diode of an interface device corresponding thereto, according to a third alternative embodiment of the invention;

45 **Figure 15** is a partially exploded schematic view of a coil loop and the output cables therefor according to a presently preferred embodiment of the invention;

50 **Figure 16** is a schematic diagram of an interface device according to another aspect of the presently preferred embodiment of the invention wherein, in its single-receiver version, the interface device has a probe interface circuit for interfacing the intracavity probe of Figure 15 with a (probe) input port of a magnetic resonance (MR) system which is not equipped with its own preamplifier;

55 **Figure 17** is a perspective view of the interface device in its single-receiver version of Figure 16, which is designed to interface the intracavity probe to the MR system via a (probe) input port thereof that is not equipped with a preamplifier;

Figure 18 is a schematic diagram of an interface device according to yet another aspect of the presently preferred

embodiment of the invention wherein, in its multiple-receiver version, the interface device has (i) a probe interface circuit for interfacing the intracavity probe of Figure 15 with a (probe) input port of an MR system equipped with its own preamplifier and (ii) an array interface circuit for interfacing a coil system, such as the Gore® torso array, with the (coil) input ports of the MR system;

Figure 19 is a perspective view of the interface device in its multiple-receiver version of Figure 18, which is designed to interface the intracavity probe and a coil system, such as the Gore® torso array, with the Phased Array Port of the MR system; and

Figures 20A and 20B illustrate top and side schematic views of an intracavity probe having two or more coil loops deployed in a phased array configuration.

DETAILED DESCRIPTION OF THE INVENTION

[0028] In all of its embodiments and related aspects, the present invention disclosed below is ideally used with magnetic resonance (MR) systems designed to operate at 3.0 Tesla field strengths, though it is also applicable to those operable from approximately 2.0 to 5.0 T. For purposes of illustration below, the invention will be described in the context of the 3.0T systems produced by General Electric Medical Systems (GEMS).

[0029] Figures 1-7 illustrate one aspect of a first embodiment of the invention, namely, an intracavity probe, generally designated 1. The probe is intended for use with an MR system to obtain images or spectra of a region of interest within a cavity of a patient. It is described herein in a specific implementation, i.e., as an endorectal probe designed to be inserted into the rectum to obtain images and/or spectra of the male prostate gland. Although presented herein as an endorectal probe, it should be understood that the invention is equally capable of being adapted to obtain images of and/or spectra from other regions of interest such as those accessible through the mouth, the vagina or other orifices penetrable by an intracavity probe. The principles presented herein may also be applied to MR imaging or spectroscopic techniques appropriate for the arteries, veins, and other structures of the body. Whatever the application, the receiving coil within intracavity probe will need to be housed in, or otherwise incorporated into, a package appropriately designed to conform to the target anatomy.

[0030] In its most novel aspects as best shown in Figure 1, intracavity probe 1 includes a coil loop 2 and an output cable 3. Ideally made of a conductive material that is flexible, coil loop 2 is preferably a single turn coil capable of picking up radio frequency (RF) signals. Designed to receive magnetic resonance RF signals from the region of interest, coil loop 2 has a plurality of capacitors including first drive capacitor 21, second drive capacitor 23, and tuning capacitor 24. The first and second drive capacitors are serially connected within coil loop 2. As is explained below, the junction node 22 at which drive capacitors 21 and 23 connect forms a virtual ground for electrically balancing and impedance matching the coil loop 2. Tuning capacitor 24 is also serially connected within coil loop 2 but diametrically opposite the junction node 22 of capacitors 21 and 23. Tuning capacitor 24 is selected to resonate coil loop 2 at an operating frequency of the MR system, which for a 3.0 Tesla scanner would be approximately 128 MHz.

[0031] Output cable 3 is designed to connect coil loop 2 to an interface device for the intracavity probe 1. Such an interface device, such as either of the ones disclosed below, at its other end in turn connects to a probe input port of the MR system 10, as shown in Figures 8 and 9. Encased within an insulating sheath, output cable 3 has a shield conductor 31 and a center conductor 32 insulatively disposed therein. The shield conductor 31 connects to the junction node 22, and the center conductor 32 connects to a node of one of the drive capacitors 21 and 23 opposite junction node 22, as shown in Figure 1. In addition, for reasons detailed below, output cable 3 preferably has an electrical length of $n(\lambda/2) + S_L$, where n is an integer, λ is the wavelength of the operating frequency of MR system 10, and S_L is a supplemental length.

[0032] Figure 2 shows the intracavity probe 1 of the present invention in fully assembled form, and Figures 3-7 illustrate various partial cross-sectional views thereof. Intracavity probe 1 includes a flexible shaft 40 and inner and outer balloons 50 and 60. The shaft 40 has a distal end whose tip 41 is preferably substantially more flexible than the remainder of the shaft, and indeed may be bonded thereto as indicated at reference numeral 15. The use of such a flexible tip 41 will reduce not only the discomfort felt by the patient but also the likelihood of perforating nearby tissue during use of the probe.

[0033] Inner balloon 50 connects to the distal end of shaft 40 and encloses tip 41 thereof, as best shown in Figure 3. Inner balloon 50 is generally cylindrical in shape except for a substantially planar section on its anterior surface 51. It can be anchored to shaft 40 by a clamp 16 and by an interference fit with the distal end of shaft 40. Coil loop 2 itself is preferably encased within 5K volt insulation over which shrink wrap or similar tubing is used, thus providing a double layer of insulation. A non-stretchable material 55, which is preferably composed of an adhesive-backed cloth, can then be used to attach the coil loop 2 to the anterior surface 51 of inner balloon 50, thus securing the coil loop 2 between the inner and outer balloons 50 and 60.

[0034] Outer balloon 60 also connects to the distal end of shaft 40, enclosing both coil loop 2 and inner balloon 50. It can be anchored to shaft 40 by a clamp 17 and by an interference fit with the distal end. Outer balloon 60 has anterior

and posterior surfaces 61 and 62. The anterior surface 61 is preferably saddle-shaped to conformably fit against a correspondingly-shaped interior surface/contour of the cavity, which in the case of the prostate probe will be the rectal prostatic bulge inferior to the ampulla of the rectum. The posterior surface 62 features at least one pair of undulating folds 63 projecting therefrom. As described below, these folds 63 enable the outer balloon 60 to properly position the coil loop 2 in operative proximity to the rectal prostatic bulge of the patient when the inner balloon 50 is inflated, which optimizes the coupling between coil loop 2 and the target anatomy. In addition, as shown in Figure 5, lateral indentations 64 are preferably provided within outer balloon 60 intermediate the anterior and posterior surfaces 61 and 62. These indentations 64 essentially form a shelf on which the sides of coil loop 2 rest during assembly of the probe 1. They essentially serve as a means of positioning the coil loop between those surfaces when balloons 50 and 60 are in the uninflated state. The balloons 50 and 60 are each preferably made of a medical-grade latex or other appropriate elastomeric material. Such material should, of course, be non-paramagnetic and exhibit low dielectric losses.

[0035] Flexible shaft 40 defines two lumens 42 and 44, as best shown in Figures 3, 4, 5 and 7. Within its cylindrical wall near its distal end, shaft 40 also defines a hole 43 in communication with lumen 42, as shown in Figure 7. Lumen 42 and hole 43 together serve as a passageway for the fluid (e.g., gas or liquid) pumped into and expelled out of inner balloon 50 when inflated and deflated, respectively. Further away from its distal end, shaft 40 defines another hole 45 in its cylindrical wall. Lumen 44 and hole 45 act as the conduit through which output cable 3 is routed from coil loop 2. Output cable 3, as shown in Figure 2, has a plug 35 at its proximal end to connect the intracavity probe 1 with the appropriate interface device.

[0036] Intracavity probe 1 further includes an anti-migration disc 46, an introducer 48, and a handle 49. Fixed to the proximal end of shaft 40, the handle 49 enables the probe 1 to be easily manipulated as its distal end, inclusive of outer balloon 60 secured thereon, is inserted into the rectum and appropriately aligned within the cavity as described below. The introducer 48, also referred to as a dilator element, is designed to be easily slid over the entire length of shaft 40. Preferably funnel-shaped, the introducer 48 can be used to manually dilate the anal sphincter to allow outer balloon 60 to be easily positioned within the cavity. Without introducer 48, the anal sphincter would contract around shaft 40 and interfere with the ability to rotationally and longitudinally position the intracavity probe 1 within the cavity. The anti-migration disc 46, composed of a semi-rigid plastic or other suitable polymer, is preferably semi-spherical in shape. As shown in Figures 2 and 4, the disc 46 defines a slot 47. This slot allows the disc 46 to be snapped onto shaft 40. When affixed to shaft 40 adjacent the anal sphincter after the probe has been inserted into the rectum, the anti-migration disc 46 prevents the probe 1 from migrating superiorly due to the normal peristaltic activity of the colon.

[0037] Intracavity probe 1 also includes a means for controlling inflation of inner balloon 50. The inflation control means preferably takes the form of a compressible inflator cuff 70, a tube 71, and a stop cock 72. A syringe of suitable size could be used in lieu of cuff 70. Tube 71 connects the inflator cuff 70, or syringe, to the lumen 42 at the proximal end of shaft 40. The stop cock 72 is connected in series with tube 71 and serves to control whether air is pumped to or released from inner balloon 50. The probe 1 also preferably features a scale 14 printed on an outer surface of shaft 40. Scale 14 provides an indication of not only the distance that shaft 40 has been inserted into the cavity but also the rotational orientation of the distal end for proper alignment of the saddle-shaped anterior surface 61 of outer balloon 60 with the prostate.

[0038] In operation, the distal end of intracavity probe 1 is inserted into the cavity via the rectum while inner balloon 50, and outer balloon 60 surrounding it, are in the uninflated state. Once the distal end is inserted, the introducer 48 can be used to keep the anal sphincter dilated and thereby enable shaft 40, and its balloon-enclosed distal end, to be easily manipulated within the cavity. With the distal end inserted and the introducer 48 in place, the scale 14 on shaft 40 can then serve as a guide to enable the clinician or other medical personnel to more accurately position the probe both rotationally and longitudinally within the cavity adjacent the region of interest. Once the intracavity probe 1 is correctly positioned, the introducer 48 can be pulled inferiorly along the shaft, thereby allowing the sphincter to contract around shaft 40. This contraction assists in holding the intracavity probe 1 in place. The anti-migration disc 46 can then be snapped onto the shaft 40 adjacent the sphincter to assure that the intracavity probe 1 stays in position during the MR scanning procedure.

[0039] Before inflating the balloons, the stop cock 72 must be switched to the open state. By pumping inflator cuff 70, the inner balloon 50 will inflate via tube 71, stop cock 72, and lumen 42 and hole 43 in shaft 40. As the inner balloon inflates, the non-stretchable material 55 that secures coil loop 2 to the anterior surface 51 of inner balloon 50 also focuses inflation of the inner balloon posteriorly so as to inflate into the undulating folds 63 of outer balloon 60. As the undulating folds 63 inflate, they soon force the posterior surface 62 (i.e., folds 63) of outer balloon 60 to abut against a wall of the cavity opposite the region of interest. As inner balloon 50 continues to inflate, the force of inflation is then directed towards the underside of the anterior surface 61 of outer balloon 60. The anterior surface 51 of inner balloon 50, with coil loop 2 attached thereto, thus forces the saddle-shaped anterior surface 61 of outer balloon 60 against the correspondingly-shaped interior contour of the cavity, namely, the prostatic region of the rectum. Once the balloons at the distal end are fully inflated, the coil loop 2 will be situated approximate the prostate gland for optimal reception of the MR signals therefrom during the MR scanning procedure. The stop cock 72 can then be switched to the closed position, thereby

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allowing the clinician to disconnect the inflator cuff 70 without deflating the balloons 50 and 60. The intracavity probe 1 can then be connected to the appropriate interface device via the plug 35 of output cable 3.

5 [0040] When the scanning procedure is completed, the clinician need only switch the stop cock 72 to the open position to deflate inner balloon 50 and outer balloon 60 therewith. Whether or not the anti-migration disc 46 is removed from shaft 40, the balloon-enclosed distal end can then be removed from the rectum merely by gently pulling on the handle 49 of the intracavity probe 1. Alternatively, the intracavity probe 1 may employ a single balloon in lieu of the double balloon version described above. It may be composed of a single-ply medical-grade latex material or other suitable elastomeric material. In this arrangement, the balloon still connects to the distal end of flexible shaft 40, and the balloon will preferably have anterior and posterior surfaces identical to those described for the double balloon version. The coil loop 2, however, will ideally be bonded or otherwise secured to the underside of the anterior surface 61 of the balloon. 10 The coil loop 2 could also be encapsulated within the anterior surface 61 during the process of manufacturing the balloon. For example, coil loop 2 could be placed on a surface of the balloon and then the balloon could be redipped to place another ply of material over the outer surface of the balloon, thus covering coil loop 2 and creating the anterior surface 61 described above. However manufactured, when the inflatable balloon is inserted into the cavity and inflated, the undulating folds 63 will press against the wall of the cavity opposite the region of interest. Upon full inflation of the balloon, its anterior surface 61 will then be forced against the correspondingly-shaped interior contour of the cavity thereby bringing the coil loop 2 into operative proximity with the region of interest (i.e., the prostate gland) wherefrom it can best receive the MR signals. 15

[0041] The disclosure further provides a method of designing the intracavity probe 1. Variations on this method, which will become apparent to skilled artisans upon reading this document, are also contemplated by the present disclosure. The first step of the method involves choosing the size of a loop of wire that will form the basis for coil loop 2. For an intracavity probe designed for imaging the prostate, the wire loop should be sized so that the distal end of the probe, inclusive of the two balloons between which coil loop 2 will be situated, can be inserted into the rectum with minimal discomfort to the patient. The next steps involve temporarily inserting a variable capacitor within the wire loop, and then 20 subjecting the loop to the operating frequency of the MR system 10. For 3.0 Tesla scanners for which the present invention is particularly well suited, the operating frequency would be approximately 128 MHz. For the GEMS 3.0T Signa® scanner, the operating frequency is actually closer to 127.74 MHz. For the Siemens 3.0T scanner, the operating frequency is 123.2MHz. 25

[0042] While the wire loop is being subjected to RF energy at the designated operating frequency, the variable capacitor should be adjusted to a value, hereinafter referred to as C_{RV} , at which the wire loop will resonate. Once resonance is achieved, the capacitive and inductive reactances of the wire loop will, of course, be equal in magnitude at the operating frequency. For the purposes of the following calculations, 10 picofarads (pF) is an ideal value for C_{RV} to establish resonance within the wire loop according to this method of designing the intracavity probe 1. Once C_{RV} has been established, the quality factor of the loop can be measured while the loop is operating under loaded conditions. There 30 are several known techniques for measuring the quality factor. One such technique involves making an S_{21} response measurement using two test probes and a network analyzer, with the two test probes being connected to ports 1 and 2, respectively, of the network analyzer. With the loops of the two test probes positioned at right angles to each other, the wire loop of the present invention is placed between them. This arrangement allows RF energy supplied to the loop of the first test probe to be induced within the wire loop, which in turn induces an RF signal in the loop of the second test probe. The two test probes then convey their respective RF signals to the network analyzer, which displays the resulting frequency response curve graphically in terms of amplitude versus frequency. Using the displayed signal, the quality factor can be ascertained by locating the center frequency of the frequency response curve and dividing it by the 3 dB bandwidth (i.e., the band between the 3 dB (half power) points at the high-pass and low-pass ends of the curve). For a 3.0 Tesla scanner, the quality factor of the loop will lie between 10 and 20. More typically, the quality factor of the 35 loop under loaded conditions will be: 40 45

$$Q_{Loaded} = 15 \text{ (measured)}$$

50 [0043] The next step of the method involves determining the series resistance, R_S , of the loop. The series resistance represents the equivalent resistive losses exhibited by the loop due to its presence within the cavity of the patient. R_S is thus not a physical component, only the effect the patient has on the loop. It reduces the quality of coil loop 2 by partially dissipating the energy within it. It can be calculated from the equation:

$$R_S = X_L/Q$$

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where Q is the quality factor measured above and X_L is the inductive reactance of the wire loop when loaded. As noted above, the capacitive and inductive reactances of the loop are equal in magnitude at resonance:

$$X_L = X_P$$

$$X_L = 2\pi f L_{\text{COIL}} \text{ and } X_P = 1/(2\pi f C_{\text{RV}})$$

where f is the operating frequency of the MR system 10. Consequently, the inductive reactance of the loop, X_L , can be calculated from:

$$X_L = 1/(2\pi f C_{\text{RV}}) = 1/(2\pi \times 128 \times 10^6 \times 10 \times 10^{-12}) = 124.34 \Omega.$$

Consequently, the series resistance of the loop will be:

$$R_S = X_L/Q_{\text{Loaded}} = 124.34 \Omega/15 = 8.29 \Omega.$$

[0044] The method also requires the step of matching the output impedance of intracavity probe 1 with the impedance required by the external circuit with which the intracavity probe shall interface. The external circuit can take the form of one of the interface devices disclosed herein, and will typically require an impedance of 50 Ω . Consequently, this step of the method includes devising an impedance matching network to match the impedance required by the external circuit, R_P , to the series resistance of the loop, R_S . In this impedance matching network, the quality of the series and parallel legs of the matching network, as represented by $Q_P = R_P/X_P$ and $Q_S = X_S/R_S$, are equal. Consequently, R_S and R_P are related by the equation:

$$R_P = (Q^2 + 1)R_S$$

where R_P can also be referred to as the equivalent parallel resistance. Given that the quality of the series and parallel legs of the matching network are equal, the quality of the matching network can then be calculated from:

$$Q = Q_{S,P} = (R_P/R_S - 1)^{1/2} = (50 \Omega/8.29 \Omega - 1)^{1/2} = 2.24.$$

[0045] The parallel reactance, X_P , associated with R_P in the impedance matching network can then be calculated from:

$$X_P = R_P/Q = 50 \Omega/2.24 = 22.32 \Omega.$$

[0046] The value of the matching capacitor can then determined from the parallel reactance:

$$C_P = 1/(2\pi f X_P) = 1/(2\pi \times 128 \times 10^6 \times 22.32) = 55.7 \text{ pF}.$$

[0047] Another step involves inserting two capacitors of the matching value into the wire loop in series with each other. These are the two drive capacitors, C_{D1} and C_{D2} , respectively designated 21 and 23, shown in Figure 1. Using the above calculations, the drive capacitors 21 and 23 together have an effective value of 27.85 pF. Junction node 22 is formed at the site at which drive capacitors 21 and 23 connect. The shield conductor 31 of output cable 3 connects to junction node 22, and the center conductor 32 connects to the node on the other side of either drive capacitor 21 or drive capacitor 23. Therefore, according to the above calculations, the value of drive capacitor 21, C_{D1} , is thus what enables the coil loop 2 to appear as a 50 ohm source to the interface device or other external circuit. This allows a 50 ohm coaxial cable to be used as the output cable 3.

[0048] A further step involves selecting a tuning capacitor, C_{TUN} , so that the total capacitance of the wire loop is equal

to the resonance value, C_{RV} . The total capacitance of the wire loop, C_{RV} , can be determined from:

$$1/C_{RV} = 1/C_{TUN} + 1/C_{D1} + 1/C_{D2} = 1/C_{TUN} + 2/C_D$$

where $C_D = C_{D1} = C_{D2}$. The value of the tuning capacitor, C_{TUN} , can then be calculated as follows:

$$\begin{aligned} C_{TUN} &= (C_{RV} * C_D) / (C_D - 2C_{RV}) \\ &= (10 \times 10^{-12} \text{ F} \times 55.7 \times 10^{-12} \text{ F}) / (55.7 \times 10^{-12} \text{ F} - 2 \times 10 \times 10^{-12} \text{ F}) \\ &= 15.6 \text{ pF} \end{aligned}$$

[0049] The variable capacitor is then removed from the wire loop and replaced with the tuning capacitor, C_{TUN} . Designated as C_T in Figure 1, the tuning capacitor 24 is serially connected within the wire loop diametrically opposite the junction node 22. Junction node 22 thus forms a virtual ground for electrically balancing the coil loop, as the electric field there is effectively zero and the voltage drop across each drive capacitor is equal but opposite in sign. This configuration results in symmetry of the electric fields relative to the patient during the receive cycle of MR system 10. It renders coil loop 2 particularly sensitive to the magnetic field, but not the electric field, components of the MR signals emitted by the region of interest. It thus enables coil loop 2 to receive the MR signals with a greater signal-to-noise ratio than prior art probes. It also does so with greater safety, as the voltages induced in the coil loop will be equal and half of what they would otherwise be if the coil loop were totally unbalanced.

[0050] Due to the high operating frequency (e.g., 128 MHz for a 3.0T MR system) and very low operating Q (i.e., between 10-20) of coil loop 2, there is no need to tune coil loop 2 on a per patient or per-coil basis, unlike the probe disclosed in U.S. Patents 5,476,095 and 5,355,087. On the basis of the above calculations including the quality factor of the loaded coil loop, the bandwidth of coil loop 2 will nominally be +/- 4.25 MHz. Consequently, assuming the coil loop will be built with +/- 2 % components, the tuning shift from probe to probe should be a maximum of approximately +/- 1.85 MHz, which is substantially less than the 3 dB bandwidth of the coil loop even without the effects of the low input impedance preamplifier explained below. The tuning is essentially fixed without material compromise due to the low Q of coil loop 2 under loaded conditions.

[0051] In this embodiment, output cable 3 preferably has an electrical length of $n(\lambda/2) + S_L$, where n is an integer, λ is the wavelength of the operating frequency of MR system 10, and S_L is a supplemental length. As best shown in Figure 1, the entire length of output cable 3 extends from coil loop 2 to its plug 35. Plug 35 represents the point at which the output cable connects to the PIN diode 33, also referred to as the decoupling diode, of the interface device or other external circuit. The $n(\lambda/2)$ part yields a section whose length is one-half the operating wavelength, which will effectively appear as zero electrical length. The value of n will typically need only be equal to 1, as coil loop 2 will in practice always be reasonably close to the circuit to which it will connect S_L represents an additional section of output cable 3 whose inductive reactance is ideally equal in magnitude to the capacitive reactance of first capacitor 21 across which the terminals of cable 3 connect. The net effect is that the entire length of output cable 3 exhibits an inductive reactance equal to the capacitive reactance of first capacitor 21.

[0052] Supplement length S_L thus inherently acts as an inductor, hereinafter referred to as L_D , which affects the operation of intracavity probe 1. During the transmit cycle of MR system 10, the MR system will decouple coil loop 2 of intracavity probe 1 from the MR system by forward biasing PIN diode 33 with a 200 mA current (see, e.g., Figure 8). This will effectively short circuit PIN diode 33, and leave the inherent inductor, L_D , of output cable 3 and first drive capacitor 21, C_{D1} , as a parallel resonant circuit. The high impedance of this parallel resonant circuit approximates an open circuit, which effectively opens coil loop 2 and thus decouples the intracavity probe 1 from the probe input port of the host MR system 10. Conversely, during the receive cycle, the MR system will couple the intracavity probe 1 to the MR system by reverse biasing decoupling diode 33 with -5V DC. This will effectively cause output cable 3 to act as a 50 ohm transmission line rather than an inductor, L_D . This will allow coil loop 2 to detect the MR signals generated within the region of interest by the resonance-inducing RF pulse transmitted by the body coil of MR system 10 (or other external coil). The MR signals will then be passed to the interface device via the conductors of cable 3.

[0053] Drive capacitors C_{D1} and C_{D2} will typically have values in the range of approximately 62 pF to 82 pF. Similarly, tuning capacitor 24, C_{TUN} , will preferably be in the range of approximately 12 to 15 pF. Better decoupling (higher open-circuit impedance) during the transmit cycle can be obtained using a value for C_{D1} in the lower end of the preferred range. Such a lower value for drive capacitor 21 would then also increase the source impedance that coil loop 2 presents to the interface device during the receive cycle. Furthermore, the exact length of S_L will depend on the particular coil

loop used within intracavity probe 1. For a coil loop that would be only lightly loaded during use, for example, drive capacitors of, say, 120 pF may be used, in which case S_L would be shorter. Conversely, for a coil loop that would be more heavily loaded, drive capacitors of 40 pF might be used, in which case S_L would be longer.

[0054] The intracavity probe 1 described above is particularly well suited for use as an endorectal coil probe with the 3.0T MR systems produced by GEMS, although it should be understood that the probe can be used for other applications as well.

[0055] Figures 8 and 9 depict two other aspects of the first embodiment of the invention, both of which designed to interface intracavity probe 1 with the GEMS MR system. In its first aspect, the interface device interfaces the intracavity probe with one receiver of the MR system, and is thus referred to as the single-receiver version. In its second aspect, the interface device interfaces both intracavity probe 1 and an external coil to the MR system using multiple receivers, and is referred to as the multiple-receiver version. As is well known, the typical GEMS Signa[®] system features four receivers and eight input ports. Receiver 0 can connect to Ports 1 or 5, receiver 1 to Ports 2 or 6, receiver 2 to Ports 3 or 7, and receiver 3 to Ports 4 or 8. In the standard configuration, the GEMS MR system has a preamplifier in each input port, except for Ports 1 and 8.

[0056] Figures 8 and 10 illustrate the interface device, generally designated 100, according to its single-receiver version. By its connector 102, interface device 100 is designed to interconnect intracavity probe 1 via output cable 3 to Port 1 of the host MR system 10, which is not equipped with its own preamplifier. Consequently, interface device 100 includes PIN diode 33 and a preamplifier 101. PIN diode 33 is connected across the input socket 103 of interface device 100 into which plug 35 of output cable 3 plugs. This design choice allows PIN diode 33 to be physically remote from the intracavity probe 1, thus allowing it to be reused as part of the interface device after the probe 1 is disposed. The preamplifier includes a GASFET 110 and a series resonant input circuit 130. The series resonant circuit 130 includes an input capacitor C_P and an input inductor L_P at the junction of which the gate of GASFET 110 is also connected. The GASFET has its source connected to biasing resistor R_B and its drain linked to coupling capacitor C_C and an RF choke RFC_2 . According well known circuit design principles, resistor R_B should be selected so that the current flowing through GASFET 110 will provide a good gain and a low noise figure. RFC_2 allows DC power to be fed to the drain of GASFET 110 without shorting out the MR RF signals output by preamplifier 101 during the receive cycle of MR system 10. A cable trap 115 is preferably employed on the other side of capacitor C_C to block undesirable cable currents.

[0057] When interface device 100 is connected to the MR system via probe cable 150 and connector 102, the drain is linked to Port 1 of MR system 10 via coupling capacitor C_C and cable trap 115. The drain is also linked to a DC power source in MR system 10 via the RF choke RFC_2 . Bypass capacitor C_{B2} connects between this RF choke and ground, and therefore carries any non-DC components to ground. Interface device 100 also includes a bypass capacitor C_{B1} and RF choke RFC_1 . Bypass capacitor C_{B1} connects between ground and a biasing line 121 with which MR system 10 is able to bias PIN diode 33. C_{B1} thus serves to carry any non-DC components away from the biasing line and decoupling diode 33. RFC_1 connects between the anode of PIN diode 33 and bypass capacitor C_{B1} , and thus presents a high impedance to RF frequencies without appreciably limiting the flow of the biasing currents. Interface device 100 also preferably includes a preamp protection diode D_{PP} and a bypass capacitor C_{B3} . Diode D_{PP} protects preamplifier 101 during the transmit cycle of the MR system. Bypass capacitor C_{B3} connects between the anode of the preamp protection diode D_{PP} and ground. RFC_3 prevents any RF currents from preamplifier 101 from flowing to MR system 10, while allowing the flow of the biasing currents on biasing line 121.

[0058] During the transmit cycle, MR system 10 will forward bias diodes D_D and D_{PP} via biasing line 121. Situated across the connector 103 of device 100 into which plug 35 of output cable 3 plugs, PIN diode D_D will thus decouple intracavity probe 1 as explained above. Meanwhile, preamp protection diode D_{PP} will effectively short circuit the gate of GASFET 110, which prevents the transmitted RF pulse signal from damaging preamplifier 101. During the receive cycle, MR system 10 will reverse bias those diodes, effectively turning them off. The series resonant circuit 130 will provide optimum impedance to GASFET 110 when coil loop 2 is operating under loaded conditions. Coupled to the gate of GASFET 110, the series resonant circuit 130 will provide preamplifier 101 with a relatively low input impedance, which serves to broaden the frequency response of coil loop 2. This broader frequency response offsets the fixed tuning scheme, which makes the tuning of coil loop 2 far less critical when compared to the probe disclosed in U.S. Patents 5,476,095 and 5,355,087. More specifically, with coil loop 2 acting as a 50 ohm input, series resonant circuit 130 will provide a high impedance (~1000 to 2000 ohms) to GASFET 110 while appearing as a very low impedance (~1 to 5 ohms) to coil loop 2. This will effectively cause coil loop 2 to decouple somewhat, which broadens its frequency response without sacrificing the signal-to-noise ratio. Along with its series resonant input circuit 130, preamplifier 101 will thus provide gain and impedance matching between the anode of decoupling diode 33 and Port 1 so that the MR signals detected by coil loop 2 are passed to Port 1 of the MR system with enhanced signal-to-noise ratio.

[0059] Interface device 100 also preferably features circuitry 160 to prevent the MR system 10 from performing a scanning procedure when the intracavity probe 1 is not connected to the interface device. Such circuitry 160 could create a driver fault within the MR system 10 to prevent it from undertaking a scan when the probe is disconnected. An audible alarm or display 161 as part of circuitry 160 through which to notify medical personnel of such a fault is also preferable.

[0060] Figures 9 and 11 illustrate the interface device, generally designated 200, according to its multiple-receiver version. By its connector 202, interface device 200 is designed to interface not only intracavity probe 1 but also a phased array coil system 80 with the Phased Array Port of the GEMS 3.0T Signa[®] MR system. The Phased Array Port is typically composed of four ports (e.g., Ports 2, 4, 5, and 7), all of which are accessible via a single connector. The prior art Gore[®] torso array is one such phased array coil system 80 that itself can be plugged via its single connector 81 into the Phased Array Port. If the Gore[®] torso array were to be used as coil system 80, coil elements A1 and A2 of Figure 9 would be the two surface coils in the anterior paddle 82, and coil elements P1 and P2 the two surface coils in the posterior paddle 83. Those two paddles each have two coil elements whose leads are routed by means of two cables 84,85 to single connector 81. It is by connector 81 that the Gore[®] torso array 80 normally plugs into the Phased Array Port of the host MR system, with each of its four coil elements being interconnected with one of the four system ports. Interface device 200, however, when used with intracavity probe 1 and the Gore[®] torso array, will interface five coil elements (i.e., coil loop 2 and coil elements A1, A2, P1 and P2) to the four-receiver Phased Array Port of MR system 10. Interface device 200 combines the four-coil torso array with the receive-only endorectal coil 1 to enable high resolution imaging of the prostate along with phased array imaging of the pelvic region.

[0061] Interface device 200 includes a probe interface circuit 210 and an array interface circuit 240. Probe interface circuit 210 includes PIN diode 33 and a cable trap 211. PIN diode 33 is connected across the input socket 203 of device 200 into which plug 35 of output cable 3 plugs. Probe cable 213, also referred to herein as circuit length 213, is used to link the decoupling diode 33 --and therethrough coil loop 2 of intracavity probe 1-- with a first port (i.e., Port 7) of MR system 10. Cable trap 211 prevents undesired current from flowing on the shield conductor of the probe cable. As shown in Figure 9, the circuit length 213 preferably has an electrical length of $n(\lambda/2)$, where n is an integer and λ is the wavelength of the operating frequency of the MR system. This makes circuit length 213 effectively appear to have zero electrical length.

[0062] The array interface circuit 240 is used to electrically interconnect phased array coil system 80 and MR system 10. It includes first and second series resonant networks 242 and 252, two $1/4$ wavelength networks 261 and 262, and a $1/4$ wavelength combiner 271. Assuming coil system 80 takes the form of the Gore[®] torso array, series resonant network 242 will convey the MR signals from anterior coil element A1 to a second port (i.e., Port 4) of MR system 10. Similarly, the other series resonant network 252 will pass the MR signals from anterior coil element A2 to a third port (i.e., Port 2). As illustrated in Figure 9, one $1/4$ wavelength network 261 is situated to receive MR signals from posterior coil element P1, and the other $1/4$ wavelength network 262 is configured to receive MR signals from posterior coil element P2. Preferably of the Wilkinson type, the $1/4$ wavelength combiner 271 is connected to the outputs of both $1/4$ wavelength networks 261 and 262. It combines the MR signals received from those two networks and conveys the resulting MR signals to a fourth port (i.e., Port 5) of MR system 10.

[0063] The first series resonant network 242 includes capacitor C_{R1} and RF choke RFC_5 . Similarly, the second series resonant network 252 includes capacitor C_{R2} and RF choke RFC_6 . The values of C_{R1} and C_{R2} are selected so that each capacitor tunes out the inductance inherent in its respective circuit path. First and second networks 242 and 252 are thus series resonant at the operating frequency of MR system 10 (i.e., they act as if having a length of $n(\lambda/2)$ where $n = 0$). This enables coil system 80 and MR system 10 to operate electrically as if there were no length to the networks 242 and 252. In addition, RF choke RFC_5 is disposed in parallel with capacitor C_{R1} , as choke RFC_6 is with capacitor C_{R2} . This is because, along the circuit paths of series resonant networks 242 and 252, MR system 10 will convey biasing signals to the decoupling diodes in coil system 80 for anterior coil elements A1 and A2. Chokes RFC_5 and RFC_6 allow those biasing signals to pass from Ports 4 and 2 to those decoupling diodes.

[0064] Furthermore, as shown in Figure 9, the length of the circuit path from the input for coil element P1 (through network 261 and combiner 271) to Port 5 is ideally one-half the operating wavelength (i.e., $n\lambda/2$). The same length applies for the circuit path extending from the input for coil element P2 to Port 5. Consequently, these circuit paths will effectively appear as zero electrical length, which permits the beneficial effects of the low impedance preamplifier in Ports 5 to be reflected back to their respective inputs. In addition, MR system 10 conveys biasing signals to the decoupling diodes for posterior coil elements P1 and P2. An RF choke and related circuitry within combiner 271 and network 261 allow biasing signals to pass from Port 5 to the decoupling diode for coil element P2. An RF choke RFC_7 and related circuitry allow biasing signals to pass from Port 8 to the decoupling diode for coil element P1. The biasing signals for coil element P1 are sourced from Port 8 so that it is independent of that for coil element P2.

[0065] During the transmit cycle, MR system 10 will forward bias decoupling diode D_D with the decoupling voltage, which is preferably superimposed on the signal line of cable 213. Situated across the connector 203 of device 200 into which plug 35 of output cable 3 plugs, PIN diode D_D will thus decouple intracavity probe 1 as explained above. MR system 10 will also simultaneously forward bias the decoupling diodes of the four coil elements A1, A2, P1, and P2 in coil system 80. This will cause those decoupling diodes to short circuit, thereby yielding parallel resonant circuits of high impedance, which will effectively open circuit the four coil elements of coil system 80. In this manner, the host MR system 10 will thus decouple both the intracavity probe 1 and the torso array 80 from the Phased Array Port of the MR system. Conversely, during the receive cycle, MR system 10 will reverse bias PIN diode D_D of probe 1 and the decoupling diodes of coil system 80, effectively turning them off. This will couple intracavity probe 1 and torso array 80 to the Phased Array

Port. This will allow coil loop 2 and coil elements A1, A2, P1 and P2 to detect the MR signals emitted from their respective regions of interest (e.g., prostate and surrounding abdominal, thoracic and pelvic regions) in response to the resonance-inducing RF pulse(s). The MR signals will then be routed through interface device 200 in the aforementioned manner and passed via connector 202 to the Phased Array Port of the host MR system 10.

[0066] Interface device 200 also preferably features circuitry 280 to prevent the MR system from performing a scanning procedure when the intracavity probe 1 is not connected to the interface device. Such circuitry 280 could include a probe sense line connected to the socket 203 into which the plug 35 of intracavity probe 1 plugs. When the probe 1 is connected to interface device 200 (i.e., plug 35 inserted into socket 203), the probe sense line would be grounded. Circuitry 280 would then detect the ground and pass an appropriate signal to Port 1 to enable the MR system to begin the scanning procedure. Should the intracavity probe not be connected to the interface device, circuitry 280 would detect the resulting open circuit and respond by altering the state of Port 1 to prevent the MR system from undertaking the scan. An audible alarm or display 281 as part of circuitry 280 through which to notify the clinician of such a fault is also preferable. Various other methods of determining whether the probe is connected to the interface device are, of course, also contemplated by the present invention.

[0067] Figure 12 illustrates the intracavity probe, and the relevant part of the interface device corresponding thereto, according to a first alternative embodiment of the invention. Specifically, Figure 12 shows coil loop 2a connected through output cable 3a to the decoupling diode D_D of the interface device. Output cable 3a is unbalanced, with its shield conductor 31a connected to junction node 22a and its center conductor 32a connected to the node on the other side of drive capacitor C_{D1} . Unlike the previously disclosed first embodiment, however, output cable 3a has an electrical length of only $n(\lambda/2)$. This is because the supplemental length S_L has been incorporated within the interface device. This can be accomplished as shown in Figure 12, for example, by assuring that the electrical length from the input socket to the decoupling diode D_D is equal to S_L . When output cable 3a of the probe is plugged into the interface device, the total electrical length from the coil loop 2a to PIN diode D_D is then equal to $n(\lambda/2) + S_L$. Although this embodiment puts S_L in the interface device rather than in output cable 3a, it still allows the intracavity probe and its corresponding interface device to operate in the same manner as the first embodiment of the invention during both the transmit and receive cycles of the MR system.

[0068] Figure 13 illustrates the intracavity probe, and the relevant part of the interface device corresponding thereto, according to a second alternative embodiment of the invention. Specifically, Figure 13 shows coil loop 2b linked to the decoupling diodes D_{D1} and D_{D2} of the interface device through a balanced output cable 3b. At one end of output cable 3b, the first and second center conductors 32b and 34b are connected to the nodes on opposite sides of drive capacitors C_{D1} and C_{D2} , respectively. When plugged into the input socket of the corresponding interface device, output cable 3b at its proximal end has its first and second center conductors 32b and 34b electrically linked to the anodes of diodes D_{D1} and D_{D2} , respectively, with its shield conductor 31b grounded with the cathodes of the two decoupling diodes. Unlike the previously disclosed first embodiment, output cable 3b has an electrical length of only $n(\lambda/2)$, as S_L has again been incorporated within the interface device. Such use of a balanced output cable 3b allows better decoupling (e.g., $2 \times 1500 \Omega$ across each drive capacitor) than the unbalanced output cable 3a used in the first alternative embodiment.

[0069] Figure 14 illustrates the intracavity probe, and the relevant part of the interface device corresponding thereto, according to a third alternative embodiment of the invention. The coil loop 2c of the probe is linked to the decoupling diode D_D of the interface device through a balanced output cable 3c. Unlike the previous embodiments, coil loop 2c is constructed with only one drive capacitor C_D , with the tuning capacitor C_T positioned within the wire loop diametrically opposite it. The values of drive capacitor C_D and tuning capacitor C_T can be calculated generally according to the foregoing method so as to enable the coil loop 2c not only to appear as a 50 ohm source to the interface device but also to resonate at the operating frequency of the MR system. At one end of output cable 3c, the first and second center conductors 32c and 34c are connected across drive capacitor C_D . When plugged into the input socket of the interface device, output cable 3c at its proximal end has its first and second conductors 32c and 34c electrically linked to the anode and cathode, respectively, of decoupling diode D_D and its shield conductor 31c grounded with the interface device. Unlike the previously disclosed first embodiment, output cable 3c has an electrical length of only $n(\lambda/2)$, as S_L has again been incorporated within the interface device.

[0070] Figure 15 illustrates a presently preferred embodiment of the intracavity probe, generally designated 11. This intracavity probe includes a coil loop and two output cables 3d and 3e. The coil loop is preferably the same as that disclosed in connection with the first embodiment, i.e., coil loop 2. The output cables 3d and 3e, however, are designed to interconnect coil loop 2 with an interface device in which improvements not found in previous embodiments are incorporated. Of equal importance to this embodiment is the choice of electrical length for the output cables 3d and 3e, as will be explained below. The actual length of each output cable extends from coil loop 2 to the end of plug 335. Plug 335 thus represents the point at which the output cables 3d and 3e connect to the interface device or other external circuit with which the intracavity probe 11 is to be used. As best shown in Figure 17, the output cables 3d and 3e are preferably housed in a single conduit 30 terminating in plug 335.

[0071] Relative to the coil loop, output cable 3d connects by its shield conductor 31d to junction node 22d and by its

center conductor 32d to the node on the other side of drive capacitor C_{D1} . A second cable, output cable 3e, connects by its shield conductor 31e to junction node 22d and by its center conductor 32e to the node on the other side of drive capacitor C_{D2} . As should be apparent to artisans of ordinary skill, the manner in which the output cables 3d and 3e connect to coil loop 2 dictate that the MR signals conveyed from coil loop 2 to one output cable will be $\lambda/2$ radians (180 degrees) out of phase with respect to those conveyed to the other output cable.

[0072] Output cables 3d and 3e each have an actual electrical length of $S_L + n(\lambda/4)$, with the value of n being an odd integer that is typically set to 1 for the reasons noted above. With this electrical length, the output cables are designed to be used with an interface device equipped with an electrical length of $\lambda/4$ at its input, i.e., between socket 303 and the decoupling diodes. As explained in detail below, this quarter-wavelength electrical length is incorporated into the inputs of both the single-receiver and multiple-receiver versions of the interface device.

[0073] Figures 16 and 17 illustrate the interface device, designated 300 according to its single-receiver version. As best shown in Figure 17, interface device 300 by its probe cable 350 and connector 302 is designed to interconnect the intracavity probe (via conduit 30 and plug 335) to Port 1 of the host MR system 10, which is not equipped with its own preamplifier. Consequently, interface device 300 includes a preamplifier 351 along with a phase shifting network 310 and two PIN diodes 33a and 33b.

[0074] Phase shifting network 310 features two sub-networks 320 and 330. The first sub-network 320 includes inductor L_1 and capacitor C_1 , and the second sub-network 330 includes inductor L_2 and capacitor C_2 . To enable the phase shifts (discussed below) to be well tuned, either capacitor C_1 or capacitor C_2 or both may be implemented in the form of a variable capacitor. PIN diode 33a is connected across the output of first sub-network 320, and PIN diode 33b is connected likewise across the output of second sub-network 330. Bypass capacitor C_4 connects between the cathode of diode 33a and ground. Similarly, bypass capacitor C_5 connects between the anode of diode 33b and ground. Bypass capacitors C_4 and C_5 serve to route non-DC components to ground while enabling the passage of DC current for the biasing of diodes 33a and 33b. Resistors R_1 and R_2 connect across PIN diodes 33a and 33b, respectively. The values of R_1 and R_2 are selected to assure that the voltage drops across diodes 33a and 33b are as equal as possible when the diodes are reversed biased during the receive cycle of MR system 10.

[0075] Interface device 300 also includes bypass capacitor C_{B1} and RF choke RFC_1 . Bypass capacitor C_{B1} connects between ground and the biasing line 131 with which MR system 10 is able to bias PIN diodes 33a and 33b. Capacitor C_{B1} serves to carry any non-DC components to ground and thus away from biasing line 131 and decoupling diodes 33a and 33b. RFC_1 connects between the anode of PIN diode 33a and bypass capacitor C_{B1} , and thus presents a high impedance to RF frequencies without appreciably limiting the flow of the biasing currents.

[0076] Preamplifier 351 is preferably the same as that disclosed in connection with the first embodiment, i.e., preamplifier 101 of interface device 100. Preamplifier 351 includes a GASFET 360 and a series resonant input circuit 370. The series resonant circuit 370 includes input capacitor C_P and input inductor L_P at the junction of which the gate of GASFET 360 is also connected. The GASFET has its source connected to biasing resistor R_B and its drain linked to coupling capacitor C_C and RF choke RFC_2 . Among other components, a cable trap 315 is preferably employed on the other side of capacitor C_C to block undesirable cable RF currents on the outer shield of probe cable 350.

[0077] When interface device 300 is connected to the MR system via probe cable 350 and connector 302, the drain is linked to Port 1 of MR system 10 via coupling capacitor C_C and cable trap 315. The drain is also linked to a DC power source in MR system 10 via the RF choke RFC_2 . Bypass capacitor C_{B2} connects between this RF choke and ground, and therefore carries any non-DC components to ground. RFC_2 allows DC power to be fed to the drain of GASFET 360 without shorting out the MR RF signals output by preamplifier 351 during the receive cycle of MR system 10. Interface device 300 also preferably includes preamp protection diode D_{PP} and bypass capacitor C_{B3} . PIN diode D_{PP} protects preamplifier 351 during the transmit cycle of the MR system. Bypass capacitor C_{B3} connects between the anode of the preamp protection diode D_{PP} and ground, and thus blocks DC components from reaching ground. RF choke RFC_3 connects at one end to the input side of capacitor C_P and at the other end between diode D_{PP} and bypass capacitor C_{B3} . RFC_3 thus provides an open RF circuit and a shorted DC circuit to prevent detuning of the tuned circuit implemented via inductor L_P and capacitor C_P .

[0078] When the plug 335 of conduit 30 is connected to the socket 303 of interface device 300, the output cables 3d and 3e of intracavity probe 11 connect to the first and second sub-networks 320 and 330, respectively, of interface device 300. It is through sub-networks 320 and 330 that a phase shift is implemented so that, during the receive cycle of MR system 10, the MR signals conveyed by output cable 3d from coil loop 2 combine constructively with the MR signals conveyed by output cable 3e from coil loop 2. In this regard, the first and second sub-networks 320 and 330 are implemented so as to impart a positive $\lambda/4$ (+90 degree) phase shift and a negative $\lambda/4$ (-90 degree) phase shift on the MR signals received from output cables 3d and 3e, respectively. As detailed below, the combined electrical lengths imparted by the output cables 3d and 3e of probe 11 and the sub-networks 320 and 330 of interface device 300 are critical to the proper operation of this preferred embodiment of the invention.

[0079] During the receive cycle, MR system 10 will reverse bias the diodes 33a, 33b and D_{PP} in interface device 300 via biasing line 131, effectively turning them off. By open-circuiting diodes 33a and 33b, MR system 10 makes it possible

for the MR signals detected by coil loop 2 to be conveyed to the input of preamplifier 351 via output cables 3d and 3e and sub-networks 320 and 330. Specifically, as noted above, the MR signals applied by coil loop 2 to output cable 3d are $\lambda/2$ radians (180 degrees) out of phase with respect to those applied to output cable 3e. The MR signals carried by output cable 3d and sub-network 320 then travel along a first combined electrical length of $S_L + \lambda/2$, with $S_L + \lambda/4$ due to output cable 3d and $+\lambda/4$ due to sub-network 320. Simultaneously, the MR signals carried by output cable 3e and sub-network 330 travel along a second combined electrical length of S_L , with $S_L + \lambda/4$ due to output cable 3e and $-\lambda/4$ due to sub-network 330. As a consequence of this $\lambda/2$ radian phase shift brought about by network 310, the MR signals carried on these two signal paths arrive at the junction of series resonant circuit 370 and the gate of GASFET 360 back in phase relative to each other. This enables MR signals from the two signal paths to combine constructively, with the combined MR signal driving preamplifier 351 in the same manner as that disclosed above in connection with preamplifier 101 of interface device 100. Likewise, the low impedance of preamplifier 351 due to series resonant circuit 370 is reflected back as an inductance to parallel resonate each of the drive capacitors C_{D1} and C_{D2} of coil loop 2, thus broadening the frequency response of the coil loop and providing a measure of preamplifier decoupling as well, all without sacrificing the signal-to-noise ratio. Along with its series resonant circuit 370, preamplifier 351 thus provides gain and impedance matching between the (MR signals output by intracavity probe 11 appearing at) decoupling diodes 33a/33b and Port 1 so that the MR signals detected by coil loop 2 are passed to Port 1 of the MR system with enhanced signal-to-noise ratio.

[0080] During the transmit cycle, MR system 10 will forward bias diodes 33a, 33b and D_{PP} via biasing line 131. By turning on diode D_{PP} , MR system 10 effectively short circuits the gate of GASFET 360, which prevents the transmitted RF pulse from damaging preamplifier 351. By turning on diodes 33a and 33b, MR system 10 causes a short-circuit to occur at the output of each sub-network at an electrical length of $\lambda/4$ from socket 303, which is the distance between that socket and the decoupling diodes. Due to the different phase shifts imparted by the two sub-networks 320 and 330, this results in different effective electrical lengths between the drive capacitors and their respective decoupling diodes. Specifically, the electrical length between drive capacitor C_{D1} and the short-circuit at the output of sub-network 320 is $S_L + \lambda/2$, with $S_L + \lambda/4$ due to output cable 3d and $+\lambda/4$ due to sub-network 320. Supplement length S_L , as noted above, inherently acts as the inductor L_D , and ideally has an inductive reactance whose magnitude is equal to that of the capacitive reactance of capacitor C_{D1} . The $\lambda/2$ section, however, effectively appears as zero electrical length because it is one-half the operating wavelength. The effective electrical length between capacitor C_{D1} and decoupling diode 33a is thus S_L during the transmit cycle of MR system 10. The forward biasing of decoupling diode 33a thus enables the inherent inductor L_D of output cable 3d and the drive capacitor C_{D1} of coil loop 2 to form a parallel resonant circuit. The high impedance of this parallel resonant circuit approximates an open circuit, which effectively opens coil loop 2 around the point where drive capacitor C_{D1} connects to it. Similarly, the electrical length between drive capacitor C_{D2} and the short-circuit at the output of sub-network 330 is S_L , with $S_L + \lambda/4$ due to output cable 3e and $-\lambda/4$ due to sub-network 330. Supplement length S_L thus represent the effective electrical length between capacitor C_{D2} and decoupling diode 33b during the transmit cycle. When forward biased, decoupling diode 33b enables the inherent inductor L_D of output cable 3e and the drive capacitor C_{D2} of coil loop 2 to form a parallel resonant circuit. The high impedance of this parallel resonant circuit approximates an open circuit, which effectively opens coil loop 2 around the point where drive capacitor C_{D2} connects to it. In the foregoing manner, the intracavity probe is decoupled from the transmit field of MR system 10 during the transmit cycle.

[0081] The decoupling of intracavity probe 11 can also be viewed from a different perspective. As noted above, each of the output cables 3d and 3e has an electrical length of $S_L + \lambda/4$. In each output cable, the supplemental length S_L , together with the corresponding drive capacitor of coil loop 2 to which it is connected, act as a source impedance and can be thought of as being connected to a transmission line whose length is $\lambda/2$ (i.e., the sum of the $\lambda/4$ section of the output cable and the $\lambda/4$ section of the sub-network to which it is connected). As is well known, standing waves at the resonant frequency points of a short-circuited transmission line produce an usual effect. In this case, where the length of this transmission line is effectively one half of the operating wavelength of the MR system (or some integer multiple thereof), the source will see an impedance identical to that at the end of the transmission line. This technique is sometimes referred to as a half-wavelength impedance transformation.

[0082] With the intracavity probe 11 connected to interface device 300 or any other suitable circuit, each output cable 3d and 3e is connected through its corresponding sub-network to one of the short-circuited decoupling diodes 33a and 33b, each of which effectively serving as the terminal end of a half-wavelength transmission line. During the transmit cycle of MR system 10, the S_L section of each output cable together with the drive capacitor to which it is connected will therefore see a short circuit at the exact point at which the S_L section segues into the $\frac{1}{2}$ wavelength transmission line. Consequently, the inductor L_D inherent in the S_L section of output cable 3d and the drive capacitor C_{D1} of coil loop 2 form a parallel resonant circuit. The high impedance of this parallel resonant circuit approximates an open circuit, which effectively opens coil loop 2 around the point where drive capacitor C_{D1} connects to it. Similarly, the inductor L_D inherent in the S_L section of output cable 3e effectively forms an open circuit with the drive capacitor C_{D2} of coil loop 2. As a result, when plugged into an interface device of the appropriate design, the intracavity probe 11 will have its coil loop 2 effectively decoupled from the transmit field during the transmit cycle of MR system 10.

[0083] Moreover, the intracavity probe 11 will decouple from the transmit field even while disconnected from the interface device. As noted above, each of the output cables 3d and 3e has an electrical length of $S_L + \lambda/4$. In each output cable, the supplemental length S_L , together with the corresponding drive capacitor of coil loop 2 to which it is connected, act as a source impedance and can be thought of as being connected to a transmission line whose length is the remaining $\lambda/4$ section of the output cable. As is well known, standing waves at the resonant frequency points of an open-circuited transmission line produce an usual effect. In this case, where the length of this transmission line has been chosen to be exactly one quarter of the operating wavelength of the MR system (or some integer multiple thereof), the source will see the exact opposite of the impedance at the end of the transmission line. This technique is sometimes referred to as a quarter-wavelength impedance transformation.

[0084] With the intracavity probe disconnected, each output cable 3d and 3e has an open-circuit at plug 335 (i.e., at the terminal end of the quarter-wavelength transmission line). During the transmit cycle of MR system 10, the S_L section of each output cable together with the drive capacitor to which it is connected will therefore see a short circuit at the exact point at which the S_L section segues into the $\lambda/4$ section of the output cable. Consequently, the inductor L_D inherent in the S_L section of output cable 3d and the drive capacitor C_{D1} of coil loop 2 form a parallel resonant circuit. The high impedance of this parallel resonant circuit approximates an open circuit, which effectively opens coil loop 2 around the point where drive capacitor C_{D1} connects to it. Similarly, the inductor L_D inherent in the S_L section of output cable 3e effectively forms an open circuit with the drive capacitor C_{D2} of coil loop 2. As a result, should medical personnel ever attempt to use the intracavity probe, according to this presently preferred embodiment, without plugging it into a suitable interface device, the coil loop 2 therein will nevertheless be effectively decoupled from the transmit field by each of the two output cables 3d and 3e.

[0085] Figures 18 and 19 illustrate the interface device, designated 400 in this presently preferred embodiment, according to its multiple-receiver version. As best shown in Figure 19, interface device 400 by its connector 402 is designed to interface intracavity probe 11 and a supplemental coil system, such as a torso array, with the receivers of a multi-receiver MR system. The supplemental coil system, for example, may be implemented in the form of the phased array coil system 80 described above in connection with interface device 200. As an extension of this example, Figure 18 shows interface device 400 being used to interconnect the supplemental coil system and the intracavity probe 11 with the Phased Array Port of the GEMS 3.0T Signa® MR system. Due to the similarity between interface devices 200 and 400 in these respects, neither the descriptions of the Phased Array Port and the supplemental coil system nor the details on how the interface device 400 connects the supplemental coil system to the Phased Array Port will be repeated here.

[0086] Interface device 400 includes a probe interface circuit 401 and an array interface circuit 440. As alluded to above, the array interface circuit 440 is preferably the same as the array interface circuit 240 disclosed in connection with interface device 200. It should be apparent, however, that the array interface circuit may be implemented in a variety of ways, the exact manner dependent upon design choice and system requirements. Probe interface circuit 401, however, includes a phase shifting network 410 and two PIN diodes 433a and 433b. It should be apparent that the probe interface circuit 401 need not contain a preamplifier if the port in the MR system to which probe interface circuit 401 is to be linked is equipped with one. If the port is not so equipped, then a preamplifier and associated circuitry identical or similar to that used in interface device 300 may be employed, as shown in Figure 16.

[0087] Phase shifting network 410 is preferably the same as phase shifting network 310 shown in Figure 16, as are PIN diodes 433a and 433b relative to diodes 33a and 33b of interface device 300. PIN diode 433a is connected across the output of first sub-network 420, and PIN diode 433b is connected likewise across the output of second sub-network 430. Like biasing line 131 of interface device 300, biasing line 431 allows MR system 10 to bias diodes 433a and 433b during the transmit and receive cycles of operation. As a consequence of the $\lambda/2$ radian phase shift brought about by network 410, the MR signals carried on these two signal paths arrive at node N back in phase relative to each other. Node N thus represents the point at which the MR signals from the two signal paths constructively combine. The combined MR signal is what ultimately drives the preamplifier in a first port (i.e., Port 7) of MR system 10.

[0088] Probe cable 413, also referred to herein as circuit length 413, is used to link the output of the decoupling diode 433a and 433b (i.e., node N) with the first port of MR system 10. Cable trap 411 prevents undesired current from flowing on the shield conductor of the probe cable. As shown in Figure 19, the circuit length 413 preferably has an electrical length of $n(\lambda/2)$, where n is an integer and λ is the wavelength of the operating frequency of the MR system. This makes circuit length 413 effectively appear to have zero electrical length. Also implemented with the first and second series resonant networks 242 and 252 of array interface circuit 240/440, as well as by the circuit paths between coil elements F1/P2 and Port 5, these zero electrical lengths permits the beneficial effects of the low impedance preamplifiers in the Ports of MR system 10 to be reflected back to their respective inputs.

[0089] When the plug 335 of conduit 30 is connected to the socket 403 of interface device 400, intracavity probe 11 is connected by its output cables 3d and 3e to the first and second sub-networks 420 and 430, respectively, of probe interface circuit 401. Similarly, when interface device 400 is connected to the MR system via connector 402, node N of probe interface circuit 401 is linked to Port 7 of MR system 10 via probe cable 413.

[0090] During the receive cycle, MR system 10 will reverse bias PIN diodes 433a and 433b in probe interface circuit

401 as well as the decoupling diodes in array interface circuit 440. With these diodes effectively turned off, intracavity probe 11 and torso array 80 will be effectively coupled to the Phased Array Port of MR system 10. This will allow the coil loop in intracavity probe 11 and the coil elements A1, A2, P1 and P2 in coil system 80 to detect the MR signals emitted from their respective regions of interest (e.g., prostate and surrounding abdominal, thoracic and pelvic regions) in response to the resonance-inducing RF pulse(s). Specifically, similar to phase shifting network 310 described above, the MR signals applied out of phase by coil loop 2 to output cables 3d and 3e are phase-aligned by sub-networks 420 and 430, respectively, with the constructively combined MR signals driving the preamplifier in the first port (i.e., Port 7) of MR system 10. Coincidentally, similar to array interface circuit 240 described above, the MR signals from the A1 and A2 coil elements are routed through series resonant networks 442 and 452, respectively, to the second and third ports (e.g., Ports 4 and 2) of MR system 10. Likewise, the MR signals from the P1 and P2 coil elements are routed through $\frac{1}{4}$ wavelength networks 461 and 462, respectively, to the fourth port (e.g., Port 5) of MR system 10 via the $\frac{1}{4}$ wavelength combiner 471. Although interface device 400 has been illustrated and described herein in the context of a 4-receiver MR system, it should be apparent that the invention is readily adaptable to MR systems having more or even fewer receivers.

[0091] During the transmit cycle, MR system 10 will forward bias PIN diodes 433a and 433b in probe interface circuit 401. Situated across first and second sub-networks 420 and 430, respectively, diodes 433a and 433b will thus decouple intracavity probe 11 as explained above in connection with interface device 300. MR system 10 will also simultaneously forward bias the decoupling diodes of the four coil elements A1, A2, P1, and P2 in coil system 80, as described above in connection with interface device 200. This will cause those decoupling diodes to short circuit, thereby yielding parallel resonant circuits of high impedance, which will effectively open circuit the four coil elements of coil system 80. In this manner, the host MR system 10 will thus decouple both the intracavity probe 11 and the torso array 80 from the Phased Array Port of the MR system. Finally, as explained supra in connection with interface device 300, intracavity probe 11 will also decouple from the transmit field even while disconnected from the interface device 400.

[0092] As should be apparent to persons of ordinary skill in the field of magnetic resonance imaging and spectroscopy, the intracavity probe in any of the above embodiments may be constructed with two or more coil loops arranged in a phased array configuration. In addition, two or more coil loops in a single intracavity probe can be oriented cooperatively to provide quadrature coverage of the region of interest. The output cables of such intracavity probes will have to be connected accordingly to properly link the coil loops to the appropriate interface device.

[0093] Figures 20A and 20B, for example, illustrate an intracavity probe 12 having four coil elements in a phased array configuration in which each coil element is critically overlapped by each of the other coil elements. In this particular manifestation, each coil element is shown with only one drive capacitor. The four output cables are preferably implemented with an electrical length of $S_L + \lambda/4$ in a manner similar to that disclosed in connection with intracavity probe 11.

[0094] Accordingly, to promote the progress of science and the useful arts, the inventor(s) hereby secure by Letters Patent exclusive rights to all subject matter embraced by the following claims for the time prescribed by the Patent Act.

Claims

1. An interface device (300) for interfacing an intracavity probe (11) with a magnetic resonance system (10) having a receive cycle and a transmit cycle of operation, said intracavity probe (11) having a coil loop (2) and a pair of output cables (3d,3e) terminating in a connector (335) therefor for connecting said coil loop (2) to said interface device (300), each of said output cables (3d,3e) connected at the other end thereof across a drive capacitor in said coil loop (2) and having an electrical length of $S_L + n(\lambda/4)$ wherein S_L is a supplemental length whose reactance is of a same magnitude as a reactance of said drive capacitor corresponding thereto, n is an odd integer and λ is a wavelength of the operating frequency of said magnetic resonance system (10), said interface device (300) comprising:

- (a) an input socket (303) to which said intracavity probe (11) is connectable via said connector (335) thereof;
- (b) a phase shifting network (310) linked to said input socket (303), said phase shifting network (310) having first and second sub-networks (320,330) for joining a first and a second of said output cables (3d,3e), respectively, when said connector (335) of said intracavity probe (11) and said input (303) of said interface device (300) are connected, said first and said second sub-networks (320,330) having electrical lengths of $\lambda/4$ and $-\lambda/4$, respectively, through which to impart a phase shift of $\lambda/4$ and $-\lambda/4$ on magnetic resonance signals received, respectively, from said first and said second output cables (3d,3e); and
- (c) a pair of PIN diodes (33a,33b) of which a first PIN diode (33a) is connected across an output of said first sub-network (320) and a second PIN diode (33b) is connected across an output of said second sub-network (330) thus demarcating the electrical lengths of $\lambda/4$ and $-\lambda/4$ of said first and said second sub-networks (320,330), respectively, from said input socket (303) of said interface device (300) to said PIN diode corresponding thereto,

the PIN diodes (33a,33b) being capable of being biased by the magnetic resonance system (10);
 wherein when said connector (335) of said intracavity probe (11) and said input socket (303) of said interface
 device (300) are connected, said phase shifting network (310) enables:

- 5 (i) during said receive cycle when said PIN diodes (33a,33b) are reversed biased, said coil loop (2) to be
 coupled through said output cables (3d,3e) to a probe input port of said magnetic resonance system (10)
 thus allowing said magnetic resonance signals received from each of said output cables (3d,3e) to be
 constructively combined and routed to said probe input port; and
- 10 (ii) during said transmit cycle when said PIN diodes (33a,33b) are forward biased, said coil loop (2) to be
 decoupled from a transmit field of said magnetic resonance system (10) due to an effective electrical length
 of S_L that arises (A) in the combination of the electrical length $S_L + n(\lambda/4)$ of said first output cable (3d) with
 the electrical length $\lambda/4$ of said first sub-network (320) and (B) in the combination of the electrical length
 $S_L + n(\lambda/4)$ of said second output cable (3e) with the electrical length $-\lambda/4$ of said second sub-network (330).
- 15 **2.** The interface device (300) of claim 1 wherein said phase shifting network (310) includes a preamplifier (351) for
 providing gain and impedance matching between said phase shifting network (310) and said probe input port so
 that said magnetic resonance signals received from said phase shifting network (310) are passed with enhancement
 of signal-to-noise ratio to said probe input port during said receive cycle.
- 20 **3.** The interface device (300) of claim 2 wherein said preamplifier (351) includes:
- (a) a GASFET (360) having a gate, a source and a drain; and
 (b) a series resonant circuit (370) connected between said phase shifting network (310) and said gate, said
 series resonant circuit for coupling said intracavity probe to said gate so as to broaden a frequency response
 of said coil loop (2), said series resonant circuit (370) comprising an input capacitor and an input inductor at a
 25 junction of which said gate of said GASFET (360) is connected, said series resonant circuit (370) during said
 receive cycle of said magnetic resonance system (10) for providing an optimum impedance to said GASFET
 (360) when said coil loop (2) is loaded.
- 30 **4.** The interface device (300) of claim 3 wherein said GASFET (360) at said source thereof is connected to a biasing
 resistor therefor and at said drain thereof is connected to a coupling capacitor and an RF choke; such that when
 said interface device (300) is connected to said magnetic resonance system (10) said drain connects to said probe
 input port via said coupling capacitor and to a DC power source in said magnetic resonance system (10) via said
 RF choke.
- 35 **5.** The interface device (300) of anyone of claims 1-4 wherein further including:
- (a) a probe cable (350) for connecting an output of said interface device (300) to said probe input port of said
 magnetic resonance system (10); and
 40 (b) a cable trap for preventing undesired current from flowing on a shield conductor of said probe cable (350).
- 6.** The interface device (300) of claim 2-5 further comprising a preamp protection diode for protecting said preamplifier
 (351) during said transmit cycle of said magnetic resonance system (10).
- 45 **7.** The interface device of claim 1-6 for interfacing a coil system (80) and said intracavity probe (11) with said magnetic
 resonance system (10), said interface device further comprising an array interface circuit (440) for electrically inter-
 connecting said coil system (80) and said magnetic resonance system (10).
- 8.** The interface device of claim 7 wherein said array interface circuit (440) includes:
- 50 (a) a first series resonant network (242) for conveying magnetic resonance signals from a first coil of said coil
 system (80) to a first coil input port of said magnetic resonance system (10);
 (b) a second series resonant network (252) for conveying magnetic resonance signals from a second coil of
 said coil system (80) to a second coil input port of said magnetic resonance system (10);
 55 (c) a pair of $\frac{1}{4}$ wavelength networks (461,462), one of said pair for receiving magnetic resonance signals from
 a third coil of said coil system (80) and an other of said pair for receiving magnetic resonance signals from a
 fourth coil of said coil system (80); and
 (d) a $\frac{1}{4}$ wavelength combiner (471) for combining such magnetic resonance signals received from said pair of

$\frac{1}{4}$ wavelength networks (461,462) and conveying such combined magnetic resonance signals to a third coil input port of said magnetic resonance system (10).

9. The interface device of claim 8 wherein said $\frac{1}{4}$ wavelength combiner (471) is a Wilkinson combiner.

10. The interface device of claim 8 or 9 wherein each of said first and said second series resonant networks (242,252) are series resonant at the operating frequency of said magnetic resonance system (10) to thereby render an electrical length thereof effectively zero.

11. The interface device of anyone of claims 8-10 wherein said $\frac{1}{4}$ wavelength combiner (371) and said $\frac{1}{4}$ wavelength networks (461,462) connected thereto provide for such magnetic resonance signals conveyed therethrough an electrical length of effectively zero at the operating frequency of said magnetic resonance system (10).

12. The interface device of anyone of claims 7-11 further including:

- (a) a probe cable (413) having a shield conductor and a center conductor insulatively disposed therein such that at an end of said probe cable said shield and said center conductors are connected across an output of said phase shifting network (410), said probe cable (413) having an electrical length of $n(\lambda/2)$; and
- (b) a cable trap for preventing undesired current from flowing on said shield conductor of said probe cable (413).

Patentansprüche

1. Schnittstellenvorrichtung (300) zum Bilden einer Schnittstelle zwischen einer Intrakavitätssonde (11) und einem Magnetresonanzsystem (10), das einen Empfangszyklus und einen Sendebetriebszyklus aufweist, wobei die Intrakavitätssonde (11) eine Spulenschleife (2) und ein Paar Ausgangskabel (3d, 3e) aufweist, die in einem Stecker (335) dafür zum Verbinden der Spulenschleife (2) mit der Schnittstellenvorrichtung (300) enden, wobei jedes der Ausgangskabel (3d, 3e) an dem anderen Ende davon durch einen Antriebskondensator in der Spulenschleife (2) verbunden ist und eine elektrische Länge von $S_L + n(\lambda/4)$ aufweist, wobei S_L eine zusätzliche Länge ist, deren Reaktanz eine gleiche Größe aufweist wie eine Reaktanz des ihr entsprechenden Antriebskondensators, n eine ungerade Ganzzahl ist und λ eine Wellenlänge der Betriebsfrequenz des Magnetresonanzsystems (10) ist, wobei die Schnittstellenvorrichtung (300) Folgendes aufweist:

- (a) eine Eingangsbuchse (303), mit der die Intrakavitätssonde (11) über ihren Stecker (335) verbunden werden kann;
- (b) ein Phasenverschiebungsnetzwerk (310), das mit der Eingangsbuchse (303) verbunden ist, wobei das Phasenverschiebungsnetzwerk (310) ein erstes und ein zweites Unternetzwerk (320, 330) zum Verbinden eines ersten beziehungsweise eines zweiten von den Ausgangskabeln (3d, 3e) aufweist, wenn der Stecker (335) der Intrakavitätssonde (11) und der Eingang (303) der Schnittstellenvorrichtung (300) verbunden sind, wobei das erste und das zweite Unternetzwerk (320, 330) elektrische Längen von $\lambda/4$ beziehungsweise $-\lambda/4$ aufweisen, durch die auf Magnetresonanzsignalen, die von dem ersten und dem zweiten Ausgangskabel (3d, 3e) empfangen werden, eine Phasenverschiebung von $\lambda/4$ beziehungsweise $-\lambda/4$ zu erteilen ist; und
- (c) ein Paar PIN-Dioden (33a, 33b), von denen eine erste PIN-Diode (33a) durch einen Ausgang des ersten Unternetzwerks (320) verbunden ist und eine zweite PIN-Diode (33b) durch einen Ausgang des zweiten Unternetzwerks (330) verbunden ist, wodurch die elektrischen Längen von $\lambda/4$ und $-\lambda/4$ des ersten beziehungsweise des zweiten Unternetzwerks (320, 330) von der Eingangsbuchse (303) der Schnittstellenvorrichtung (300) zu der ihr entsprechenden PIN-Diode abgegrenzt werden, wobei die PIN-Dioden (33a, 33b) in der Lage sind, durch das Magnetresonanzsystem (10) vorgespannt zu werden; wobei, wenn der Stecker (335) der Intrakavitätssonde (11) und die Eingangsbuchse (303) der Schnittstellenvorrichtung (300) verbunden werden, das Phasenverschiebungsnetzwerk (310) Folgendes ermöglicht:

- (i) während des Empfangszyklus, wenn die PIN-Dioden (33a, 33b) in Sperrrichtung vorgespannt sind, das Koppeln der Spulenschleife (2) durch die Ausgangskabel (3d, 3e) an einen Sondeneingangsport des Magnetresonanzsystems (10), wodurch es ermöglicht wird, die Magnetresonanzsignale, die von jedem der Ausgangskabel (3d, 3e) empfangen werden, konstruktiv zu kombinieren und an den Sondeneingangsport weiterzuleiten; und
- (ii) während des Sendezyklus, wenn die PIN-Dioden (33a, 33b) in Durchlassrichtung vorgespannt sind, das Entkoppeln der Spulenschleife (2) von einem Sendefeld des Magnetresonanzsystems (10) aufgrund einer

wirksamen elektrischen Länge von S_L , die (A) in der Kombination der elektrischen Länge $S_L + n(\lambda/4)$ des ersten Ausgangskabels (3d) mit der elektrischen Länge $\lambda/4$ des ersten Unternetzwerks (320) und (B) in der Kombination der elektrischen Länge $S_L + n(\lambda/4)$ des zweiten Ausgangskabels (3e) mit der elektrischen Länge $-\lambda/4$ des zweiten Unternetzwerks (330) entsteht.

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2. Schnittstellenvorrichtung (300) nach Anspruch 1, wobei das Phasenverschiebungsnetzwerk (310) einen Vorverstärker (351) aufweist, um Verstärkungs- und Impedanzanpassung zwischen dem Phasenverschiebungsnetzwerk (310) und dem Sondeneingangsport bereitzustellen, derart, dass die Magnetresonanzsignale, die von dem Phasenverschiebungsnetzwerk (310) empfangen werden, während des Empfangszyklus mit einer Verbesserung des Signal-Rausch-Verhältnisses an den Sondeneingangsport weitergeleitet werden.

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3. Schnittstellenvorrichtung (300) nach Anspruch 2, wobei der Vorverstärker (351) Folgendes aufweist:

(a) einen GASFET (360), der ein Gate, eine Source und einen Drain aufweist; und

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(b) einen Reihenresonanzkreis (370), der zwischen dem Phasenverschiebungsnetzwerk (310) und dem Gate verbunden ist, wobei der Reihenresonanzkreis zum Koppeln der Intrakavitätssonde an das Gate ausgestaltet ist, um einen Frequenzgang der Spulenschleife (2) zu verbreitern, der Reihenresonanzkreis (370) einen Eingangskondensator und einen Eingangsinduktor aufweist, von dem ein Übergang mit dem Gate des GASFET (360) verbunden ist, wobei der Reihenresonanzkreis (370) während des Empfangszyklus des Magnetresonanzsystems (10) zum Bereitstellen einer optimalen Impedanz für den GASFET (360) ausgestaltet ist, wenn die Spulenschleife (2) geladen ist.

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4. Schnittstellenvorrichtung (300) nach Anspruch 3, wobei der GASFET (360) an der Source davon mit einem Spannungswiderstand dafür verbunden ist und an dem Drain davon mit einem Kopplungskondensator und einer HF-Drossel verbunden ist; derart, dass, wenn die Schnittstellenvorrichtung (300) mit dem Magnetresonanzsystem (10) verbunden ist, der Drain sich über den Kopplungskondensator mit dem Sondeneingangsport und über die HF-Drossel mit einer Gleichstrom-Leistungsquelle in dem Magnetresonanzsystem (10) verbindet.

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5. Schnittstellenvorrichtung (300) nach einem der Ansprüche 1 bis 4, die ferner Folgendes aufweist:

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(a) ein Sondenkabel (350) zum Verbinden eines Ausgangs der Schnittstellenvorrichtung (300) mit dem Sondeneingangsport des Magnetresonanzsystems (10); und

(b) eine Kabelsperre zum Verhindern, dass ein unerwünschter Strom auf einem Schutzleiter des Sondenkabels (350) fließt.

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6. Schnittstellenvorrichtung (300) nach Anspruch 2 bis 5, die ferner eine Vorverstärker-Schutzdiode zum Schützen des Vorverstärkers (351) während des Sendezyklus des Magnetresonanzsystems (10) aufweist.

7. Schnittstellenvorrichtung nach Anspruch 1 bis 6, zum Bilden einer Schnittstelle zwischen einem Spulensystem (80) und der Intrakavitätssonde (11) und dem Magnetresonanzsystem (10), wobei die Schnittstellenvorrichtung ferner eine Array-Schnittstellenschaltung (440) zum elektrischen Zusammenschalten des Spulensystems (80) und des Magnetresonanzsystems (10) aufweist.

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8. Schnittstellenvorrichtung nach Anspruch 7, wobei die Array-Schnittstellenschaltung (440) Folgendes aufweist:

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(a) ein erstes Reihenresonanznetzwerk (242) zum Befördern von Magnetresonanzsignalen von einer ersten Spule des Spulensystems (80) zu einem ersten Spuleneingangsport des Magnetresonanzsystems (10);

(b) ein zweites Reihenresonanznetzwerk (252) zum Befördern von Magnetresonanzsignalen von einer zweiten Spule des Spulensystems (80) zu einem zweiten Spuleneingangsport des Magnetresonanzsystems (10);

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(c) ein Paar $\frac{1}{4}$ -Wellenlängen-Netzwerke (461, 462), wobei eines von dem Paar zum Empfangen von Magnetresonanzsignalen von einer dritten Spule des Spulensystems (80) und ein anderes von dem Paar zum Empfangen von Magnetresonanzsignalen von einer vierten Spule des Spulensystems (80) ausgestaltet ist; und

(d) einen $\frac{1}{4}$ -Wellenlängenkombinierer (471) zum Kombinieren solcher Magnetresonanzsignale, die von dem Paar von $\frac{1}{4}$ -Wellenlängennetzwerken (461, 462) empfangen werden, und Befördern solcher kombinierter Magnetresonanzsignale zu einem dritten Spuleneingangsport des Magnetresonanzsystems (10).

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9. Schnittstellenvorrichtung nach Anspruch 8, wobei der $\frac{1}{4}$ -Wellenlängenkombinierer (471) ein Wilkinson-Kombinierer ist.

10. Schnittstellenvorrichtung nach Anspruch 8 oder 9, wobei jeder von dem ersten und dem zweiten Reihenresonanznetzwerk (242, 252) bei der Betriebsfrequenz des Magnetresonanzsystems (10) reihenresonant ist, um dadurch eine elektrische Länge davon effektiv gleich null zu machen.

11. Schnittstellenvorrichtung nach einem der Ansprüche 8 bis 10, wobei der $\frac{1}{4}$ -Wellenlängenkombinierer (371) und die $\frac{1}{4}$ -Wellenlängen-Netzwerke (461, 462), die damit verbunden sind, dafür sorgen, dass solche Magnetresonanzsignale, die dort hindurch befördert werden, bei der Betriebsfrequenz des Magnetresonanzsystems (10) eine elektrische Länge von effektiv null aufweisen.

12. Schnittstellenvorrichtung nach einem der Ansprüche 7 bis 11, die ferner Folgendes aufweist:

(a) ein Sondenkabel (413), das einen Schutzleiter und einen Mittelleiter aufweist, die isolierend darin angeordnet sind, derart, dass an einem Ende des Sondenkabels der Schutz- und der Mittelleiter durch einen Ausgang des Phasenverschiebungsnetzwerks (410) verbunden sind, wobei das Sondenkabel (413) eine elektrische Länge von $n(\lambda/2)$ aufweist; und

(b) eine Kabelsperre zum Verhindern des Fließens eines unerwünschten Stromes auf dem Schutzleiter des Sondenkabels (413).

Revendications

1. Dispositif d'interface (300) pour interfacier une sonde intracavitaire (11) avec un système de résonance magnétique (10) ayant un cycle de réception et un cycle de transmission de fonctionnement, ladite sonde intracavitaire (11) comportant une boucle de bobine (2) et une paire de câbles de sortie (3d, 3e) se terminant par un connecteur (335) pour connecter ladite boucle de bobine (2) audit dispositif d'interface (300), chacun desdits câbles de sortie (3d, 3e) étant connecté à l'autre extrémité de ceux-ci à travers un condensateur d'attaque dans ladite boucle de bobine (2) et ayant une longueur électrique $S_L + n(\lambda/4)$, où S_L est une longueur supplémentaire dont la réactance est d'une même grandeur qu'une réactance dudit condensateur d'attaque lui correspondant, n est un entier impair et λ est une longueur d'onde de la fréquence de fonctionnement dudit système à résonance magnétique (10), ledit dispositif d'interface (300) comprenant :

(a) une prise d'entrée (303) à laquelle ladite sonde intracavitaire (11) peut être connectée par l'intermédiaire dudit connecteur (335) de celle-ci ;

(b) un réseau de déphasage (310) connecté à ladite prise d'entrée (303), ledit réseau de déphasage (310) ayant des premier et second sous-réseaux (320, 330) pour joindre un premier et un second desdits câbles de sortie (3d, 3e), respectivement, lorsque ledit connecteur (335) de ladite sonde intracavitaire (11) et ladite entrée (303) dudit dispositif d'interface (300) sont connectés, lesdits premier et second sous-réseaux (320, 330) ayant des longueurs électriques de $\lambda/4$ et $-\lambda/4$, respectivement, à travers lesquels appliquer un déphasage de $\lambda/4$ et $-\lambda/4$ sur les signaux de résonance magnétique reçus, respectivement, desdits premier et second câbles de sortie (3d, 3e) ; et

(c) une paire de diodes PIN (33a, 33b) dont une diode PIN (33a) est connectée à travers une sortie dudit premier sous-réseau (320) et une seconde diode PIN (33b) est connectée à travers une sortie dudit second sous-réseau (330), délimitant ainsi les longueurs électriques $-\lambda/4$ et $-\lambda/4$ desdits premier et second sous-réseaux (320, 330), respectivement, de ladite prise d'entrée (303) dudit dispositif d'interface (300) de ladite diode PIN lui correspondant, les diodes PIN (33a, 33b) pouvant être polarisées par le système de résonance magnétique (10) ; où ledit connecteur (335) de ladite sonde intracavitaire (11) et ladite prise d'entrée (303) dudit dispositif d'interface (300) sont connectés, ledit réseau de déphasage (310) permet :

(i) au cours dudit cycle de réception, lorsque lesdites diodes PIN (33a, 33b) sont polarisées en inverse, à ladite boucle de bobine (2) d'être couplée par l'intermédiaire desdits câbles de sortie (3d, 3e) à un port d'entrée de sonde dudit système de résonance magnétique (10), permettant ainsi auxdits signaux de résonance magnétique reçus de chacun desdits câbles de sortie (3d, 3e) d'être combinés de manière constructive et acheminés vers ledit port d'entrée de sonde ; et

(ii) au cours dudit cycle de transmission, lorsque lesdites diodes PIN (33a, 33b) sont polarisées en direct, à ladite boucle de bobine (2) d'être découplée d'un champ d'émission dudit système de résonance magnétique (10) du fait d'une longueur électrique effective de S_L qui arrive (A) dans la combinaison de la longueur électrique $S_L + n(\lambda/4)$ dudit premier câble de sortie (3d) avec la longueur électrique $\lambda/4$ dudit premier sous-réseau (320) et (B) dans la combinaison de la longueur électrique $S_L + n(\lambda/4)$ dudit second câble de sortie

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(3e) avec la longueur électrique $-\lambda/4$ dudit second sous-réseau (330).

2. Dispositif d'interface (300) selon la revendication 1, dans lequel ledit réseau de déphasage (310) comprend un préamplificateur (351) pour apporter un gain et une adaptation d'impédance entre ledit réseau de déphasage (310) et ledit port d'entrée de sonde de manière à ce que lesdits signaux de résonance magnétique reçus dudit réseau de déphasage (310) soient transmis avec une amélioration du rapport signal sur bruit audit port d'entrée de sonde au cours dudit cycle de réception.
3. Dispositif d'interface (300) selon la revendication 2, dans lequel ledit préamplificateur (351) comprend :
- (a) un GASFET (360) ayant une grille, une source et un drain ; et
 - (b) un circuit résonnant série (370) connecté entre ledit réseau de déphasage (310) et ladite grille, ledit circuit résonnant série étant destiné à coupler ladite sonde intracavitaire à ladite grille de manière à élargir une réponse en fréquence de ladite boucle de bobine (2), ledit circuit résonnant série (370) comprenant un condensateur d'entrée et une inductance d'entrée à une jonction de laquelle ladite grille dudit GASFET (360) est connectée, ledit circuit résonnant série (370), pendant ledit cycle de réception dudit système de résonance magnétique (10), procurant une impédance optimale audit GASFET (360) lorsque ladite boucle de bobine (2) est chargée.
4. Dispositif d'interface (300) selon la revendication 3, dans lequel ledit GASFET (360), au niveau de ladite source de celui-ci, est connecté à une résistance de polarisation et, au niveau dudit drain de celui-ci, est connecté à un condensateur de couplage et à une bobine d'arrêt RF ; de sorte que lorsque ledit dispositif d'interface (300) est connecté audit système de résonance magnétique (10), ledit drain se connecte audit port d'entrée de sonde par l'intermédiaire dudit condensateur de couplage et à une source d'alimentation en courant continu dans ledit système par résonance magnétique (10) par l'intermédiaire de ladite bobine d'arrêt RF.
5. Dispositif d'interface (300) selon l'une quelconque des revendications 1 à 4, comprenant en outre :
- (a) un câble de sonde (350) pour connecter une sortie dudit dispositif d'interface (300) audit port d'entrée de sonde dudit système de résonance magnétique (10) ; et
 - (b) un piège de câble pour empêcher qu'un courant indésirable ne s'écoule sur un conducteur blindé dudit câble de sonde (350).
6. Dispositif d'interface (300) selon une des revendications 2 à 5, comprenant en outre une diode de protection de préampli pour protéger ledit préamplificateur (351) au cours dudit cycle de transmission dudit système de résonance magnétique (10).
7. Dispositif d'interface selon une des revendications 1 à 6 destiné à interfacier un système de bobine (80) et ladite sonde intracavitaire (11) avec ledit système de résonance magnétique (10), ledit dispositif d'interface comprenant en outre un circuit d'interface de réseau (440) pour interconnecter électriquement lesdits système de bobine (80) et ledit système de résonance magnétique (10).
8. Dispositif d'interface selon la revendication 7, dans lequel ledit circuit d'interface de réseau (440) comprend :
- (a) un premier circuit résonnant série (242) pour acheminer des signaux de résonance magnétique d'une première bobine dudit système de bobine (80) à un premier port d'entrée de bobine dudit système de résonance magnétique (10) ;
 - (b) un second réseau résonnant série (252) pour acheminer des signaux de résonance magnétique d'une seconde bobine dudit système de bobine (80) à un second port d'entrée dudit système de résonance magnétique (10) ;
 - (c) une paire de réseaux quart d'onde (461, 462), une de ladite paire pour recevoir des signaux de résonance magnétique provenant d'une troisième bobine dudit système de bobine (80) et une autre de ladite paire pour recevoir des signaux de résonance magnétique d'une quatrième bobine dudit système de bobine (80) ; et
 - (d) un combineur quart d'onde (471) pour combiner les signaux de résonance magnétique reçus de ladite paire de réseaux quart d'onde (461, 462) et acheminer de tels signaux de résonance magnétique combinés à un troisième port d'entrée de bobine dudit système de résonance magnétique (10).
9. Dispositif d'interface selon la revendication 8, dans lequel ledit combineur quart d'onde (471) est un combineur Wilkinson.

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10. Dispositif d'interface selon la revendication 8 ou la revendication 9, dans lequel chacun desdits premier et desdits seconds réseaux résonnants en série (242, 252) sont résonnant en série à la fréquence de fonctionnement dudit système de résonance magnétique (10) pour ainsi rendre une longueur électrique de celui-ci effectivement nulle.

5 11. Dispositif d'interface selon l'une quelconque des revendications 8 à 10, dans lequel ledit combineur quart d'onde (371) et lesdits réseaux quart d'onde (461, 462) connectés à celui-ci délivrent de tels signaux de résonance magnétique acheminés à travers une longueur électrique effectivement nulle à la fréquence de fonctionnement dudit système de résonance magnétique (10).

10 12. Dispositif d'interface selon l'une quelconque des revendications 7 à 11, comprenant en outre :

(a) un câble de sonde (413) comprenant un conducteur de blindage et un conducteur central disposé dans celui-ci de manière isolante de manière à ce qu'à une extrémité dudit câble de sonde, ledit blindage et lesdits conducteurs centraux soient connectés à travers une sortie dudit réseau de déphasage (410), ledit câble de

15 sonde (413) ayant une longueur électrique de $n(\lambda/2)$; et

(b) un piège de câble pour empêcher qu'un courant indésirable ne s'écoule sur ledit conducteur blindé dudit câble de sonde (413).

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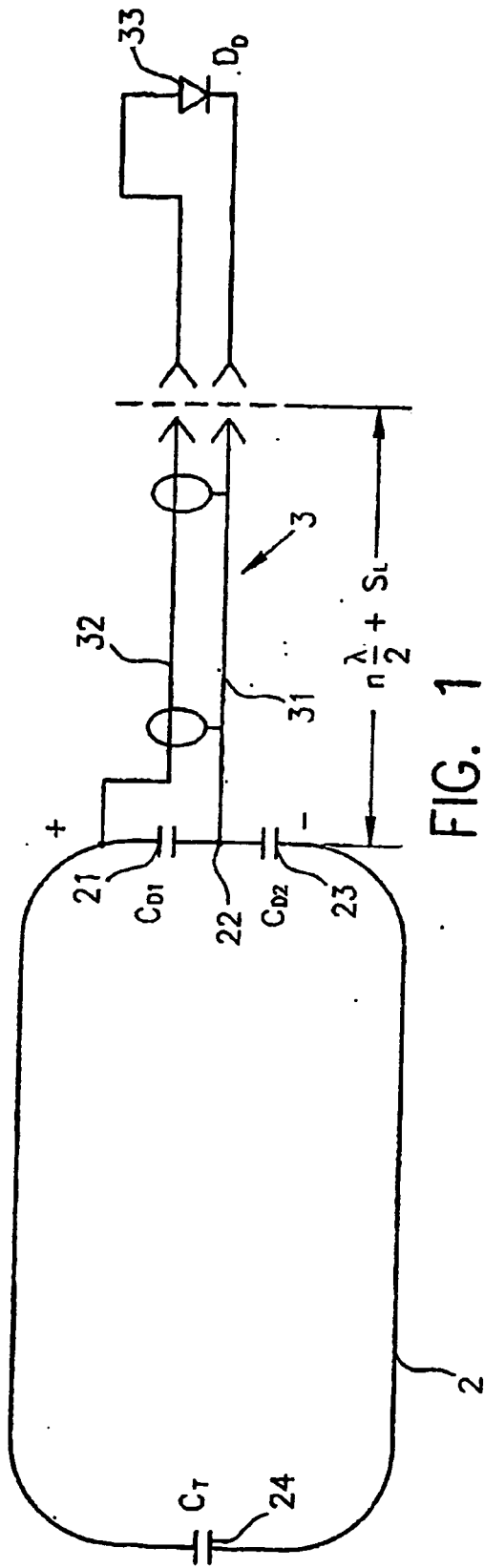


FIG. 1

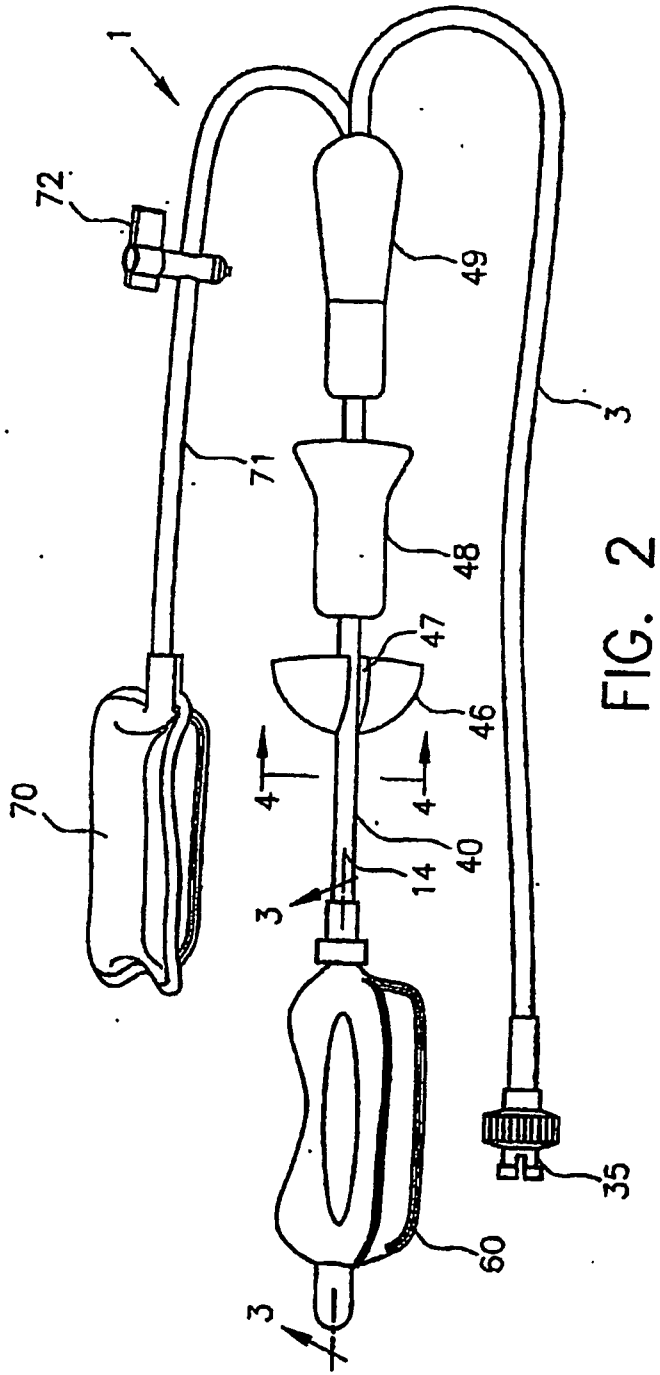


FIG. 2

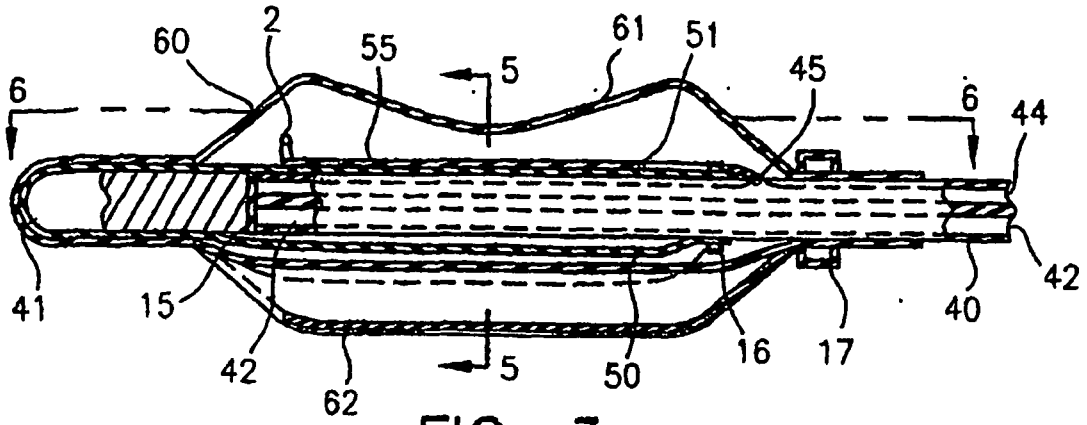


FIG. 3

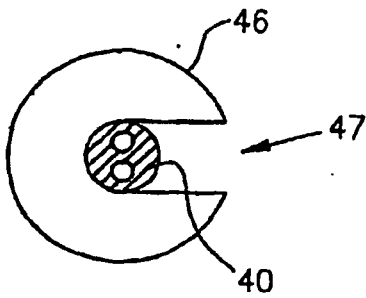


FIG. 4

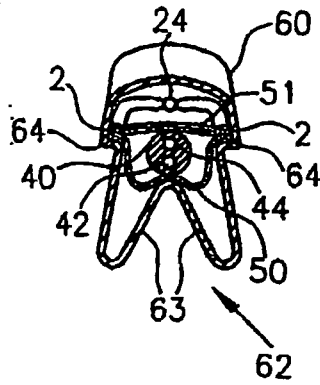


FIG. 5

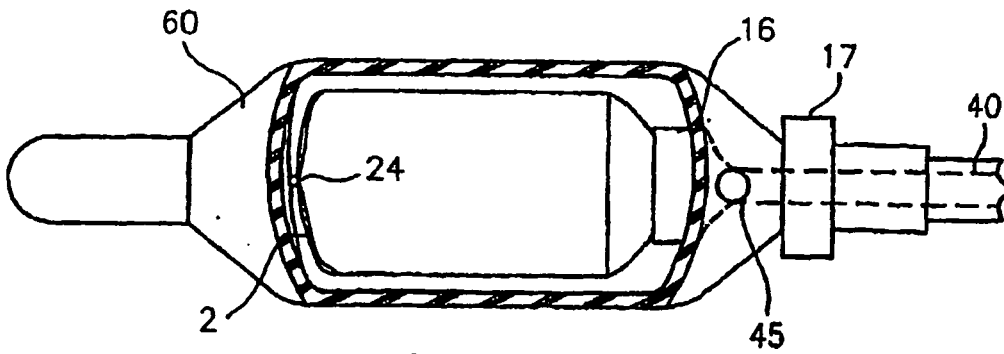


FIG. 6

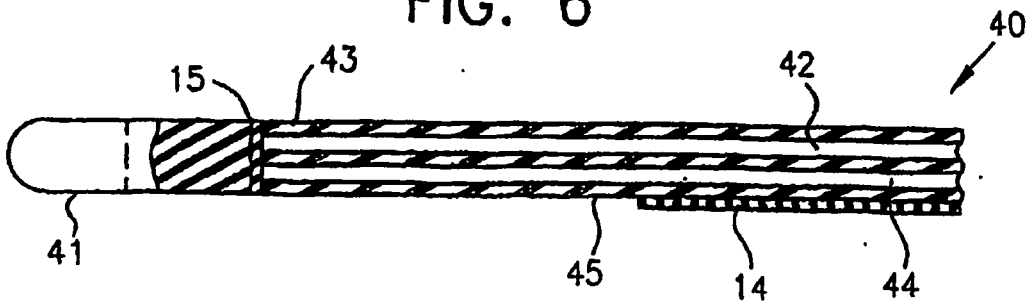


FIG. 7

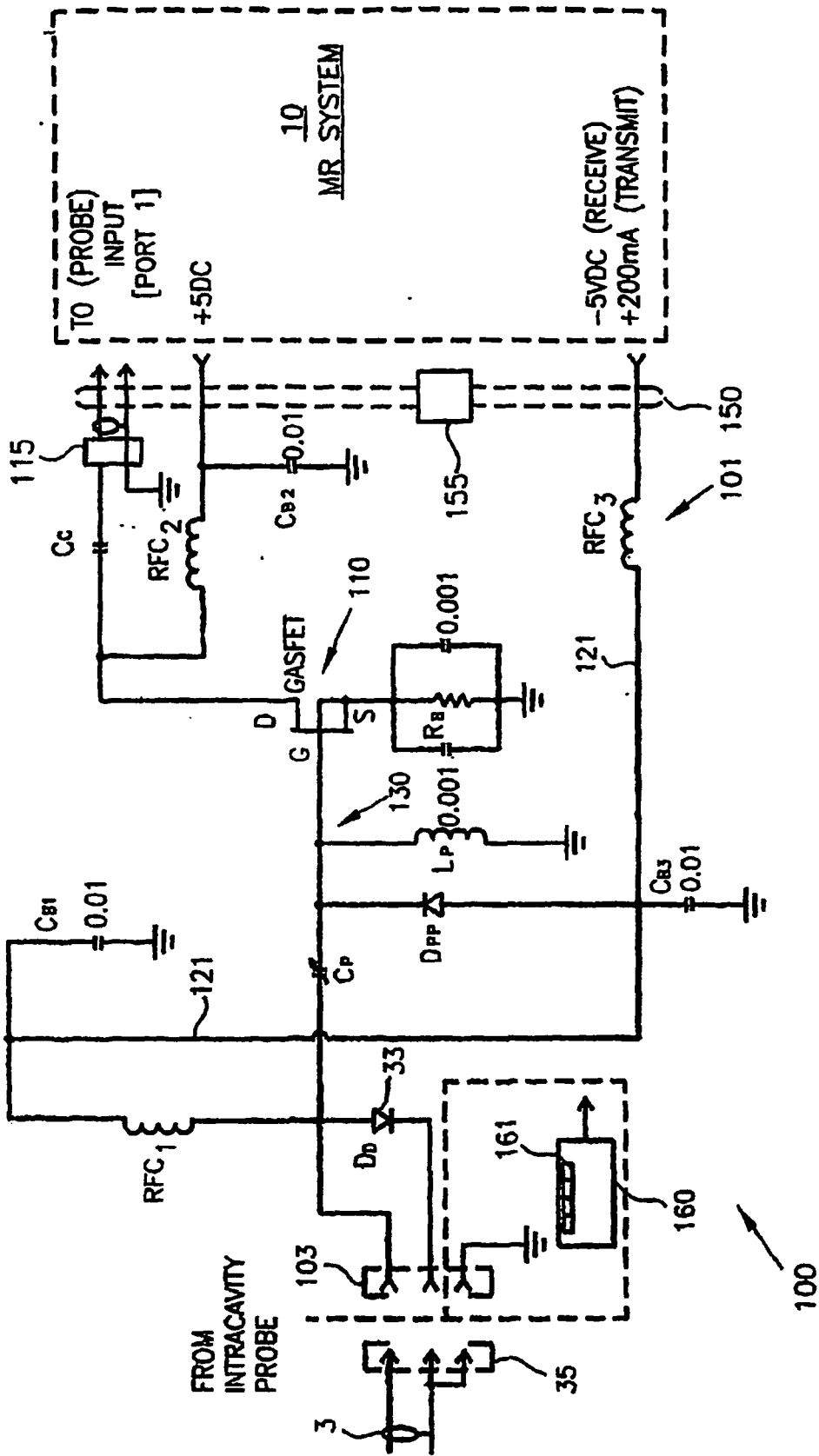


FIG. 8

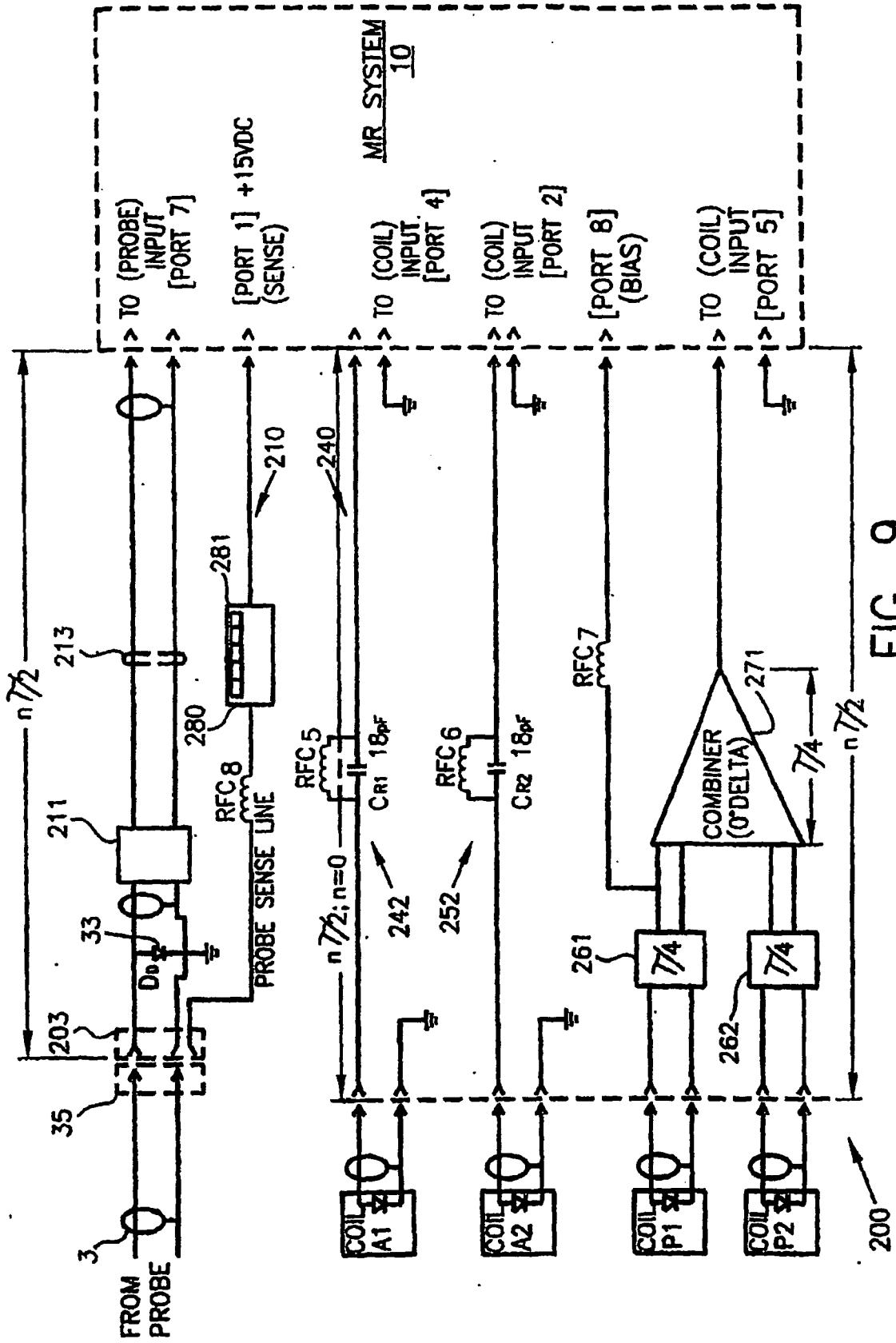


FIG. 9

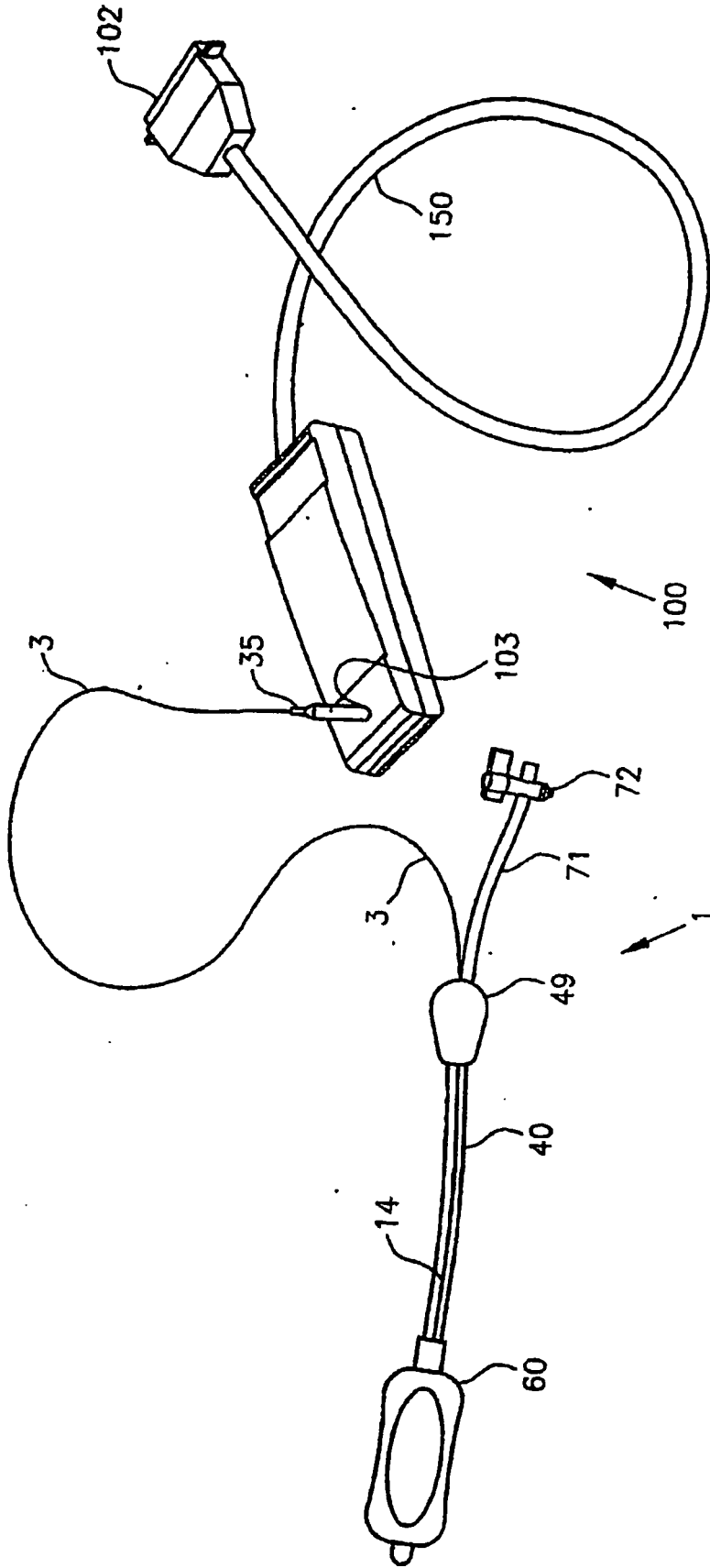


FIG. 10

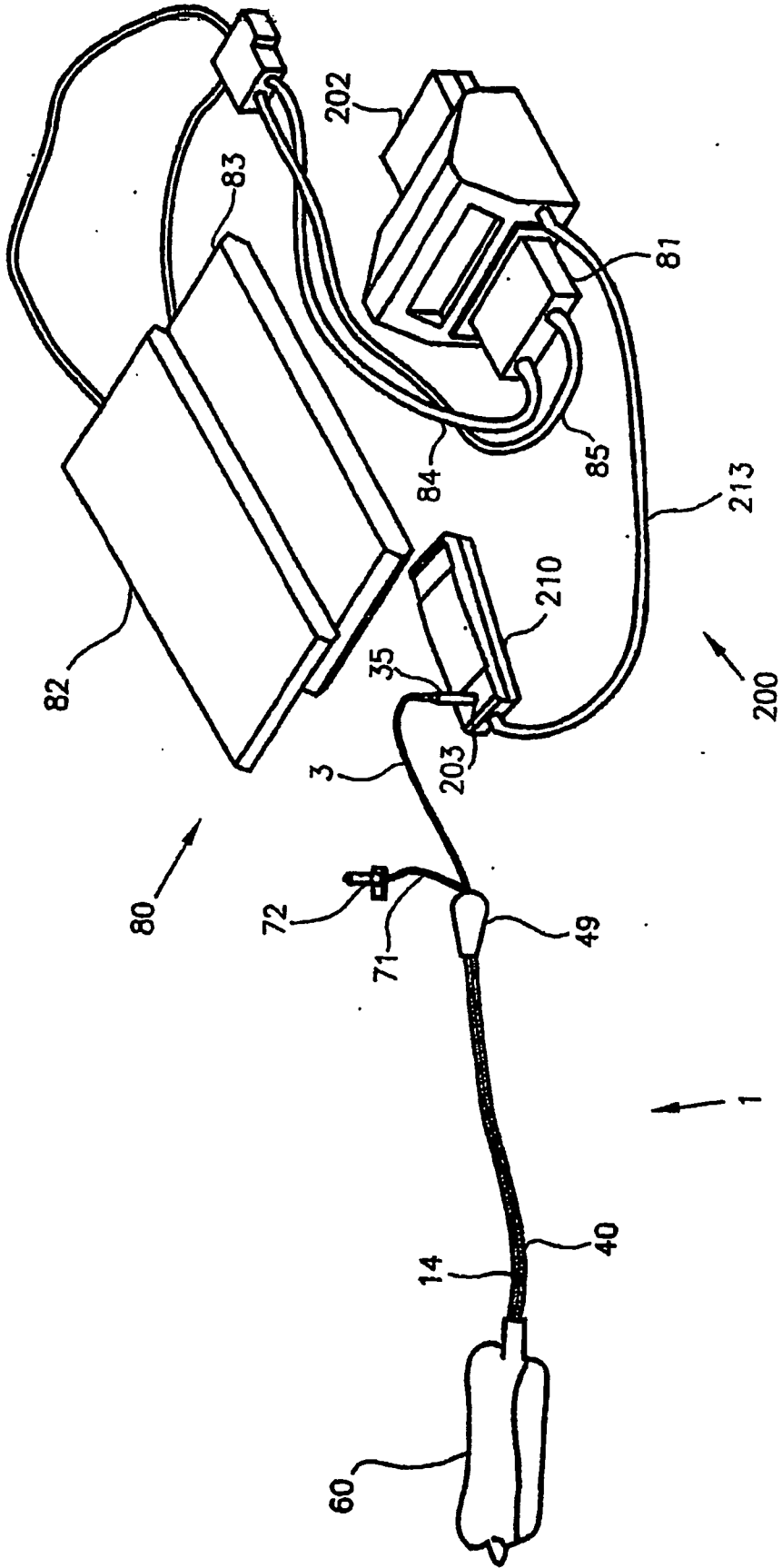


FIG. 11

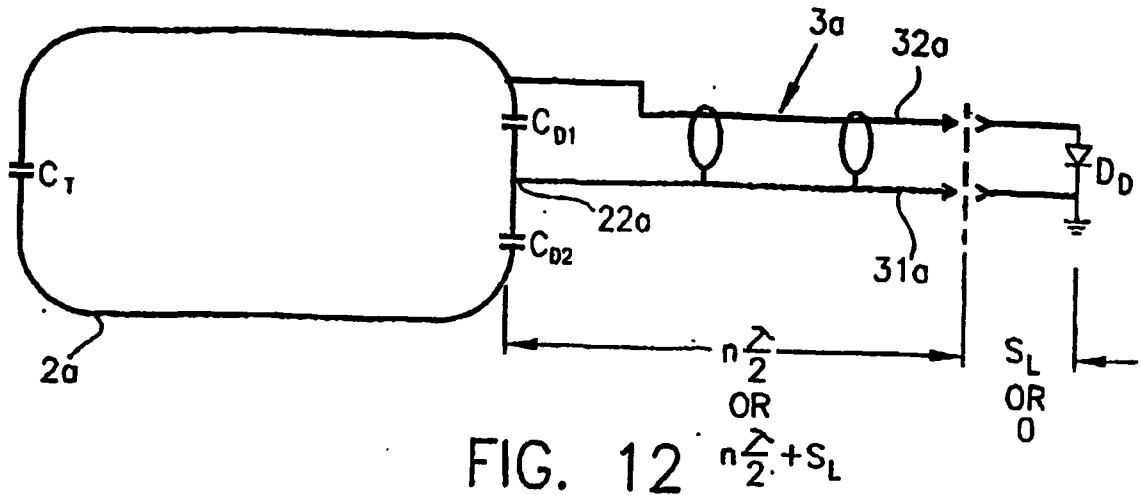


FIG. 12

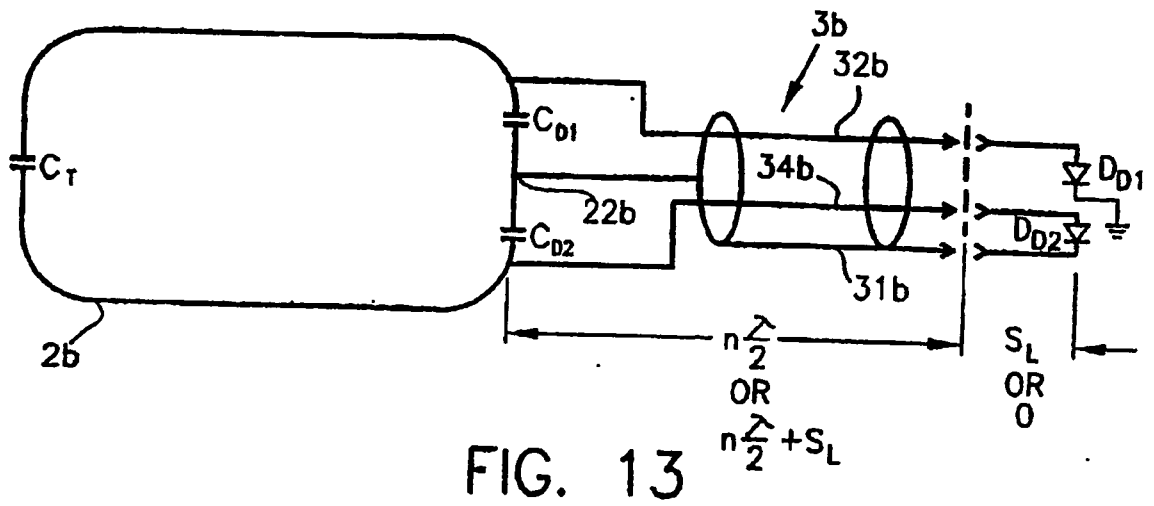


FIG. 13

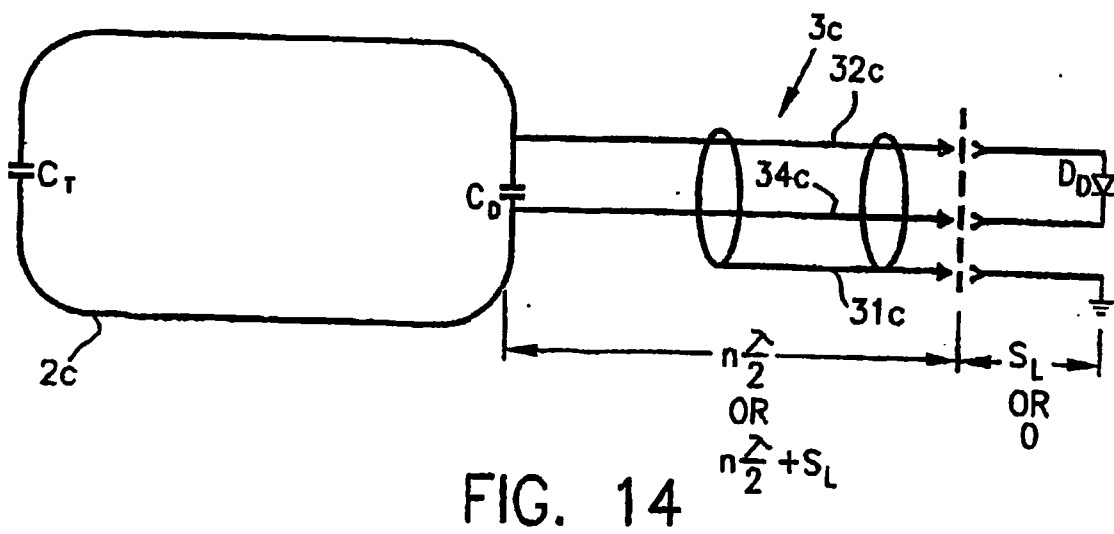


FIG. 14

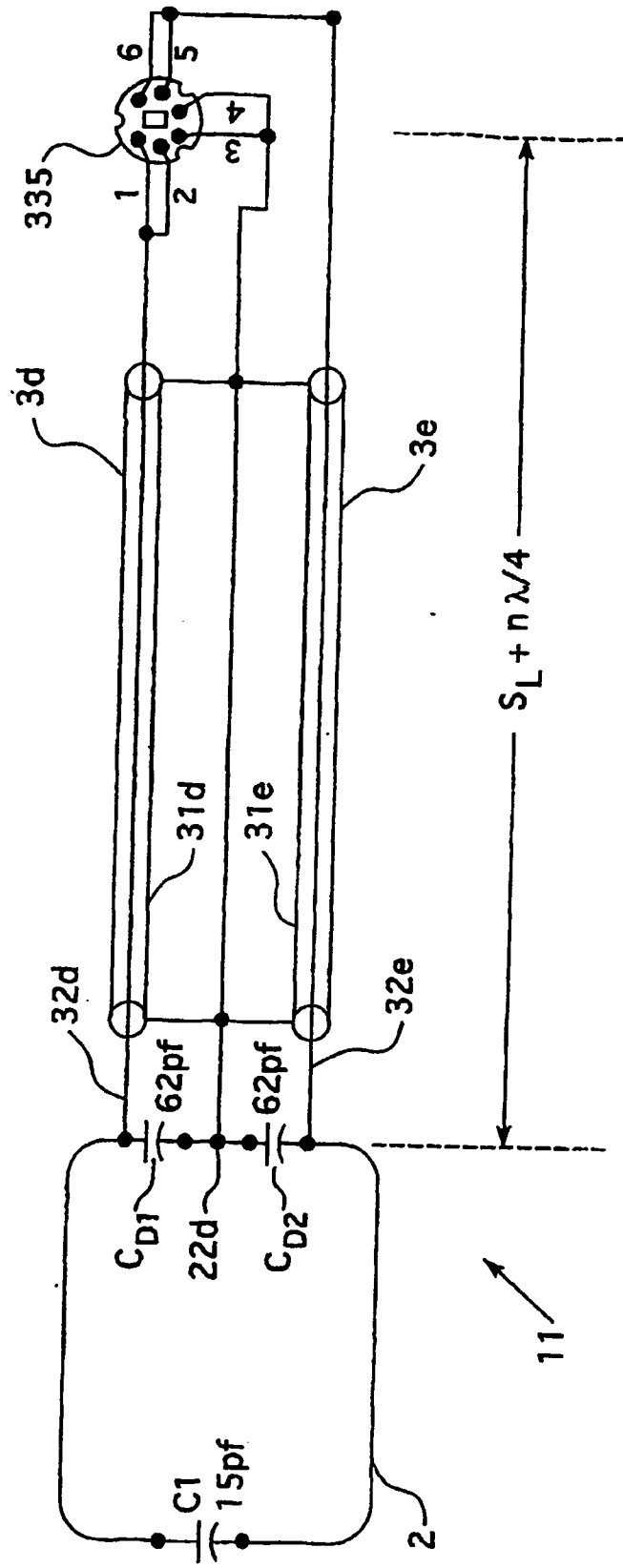
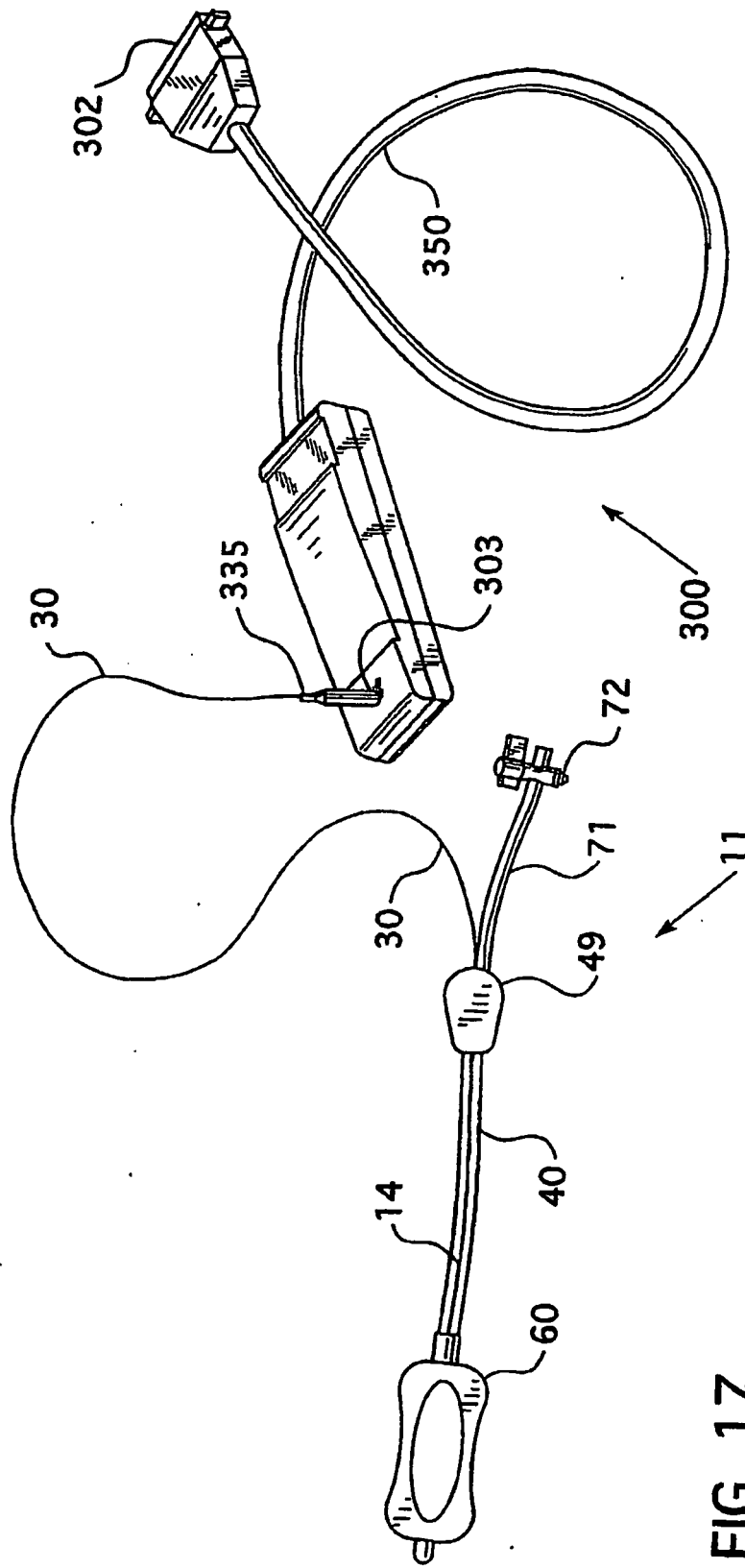


FIG. 15



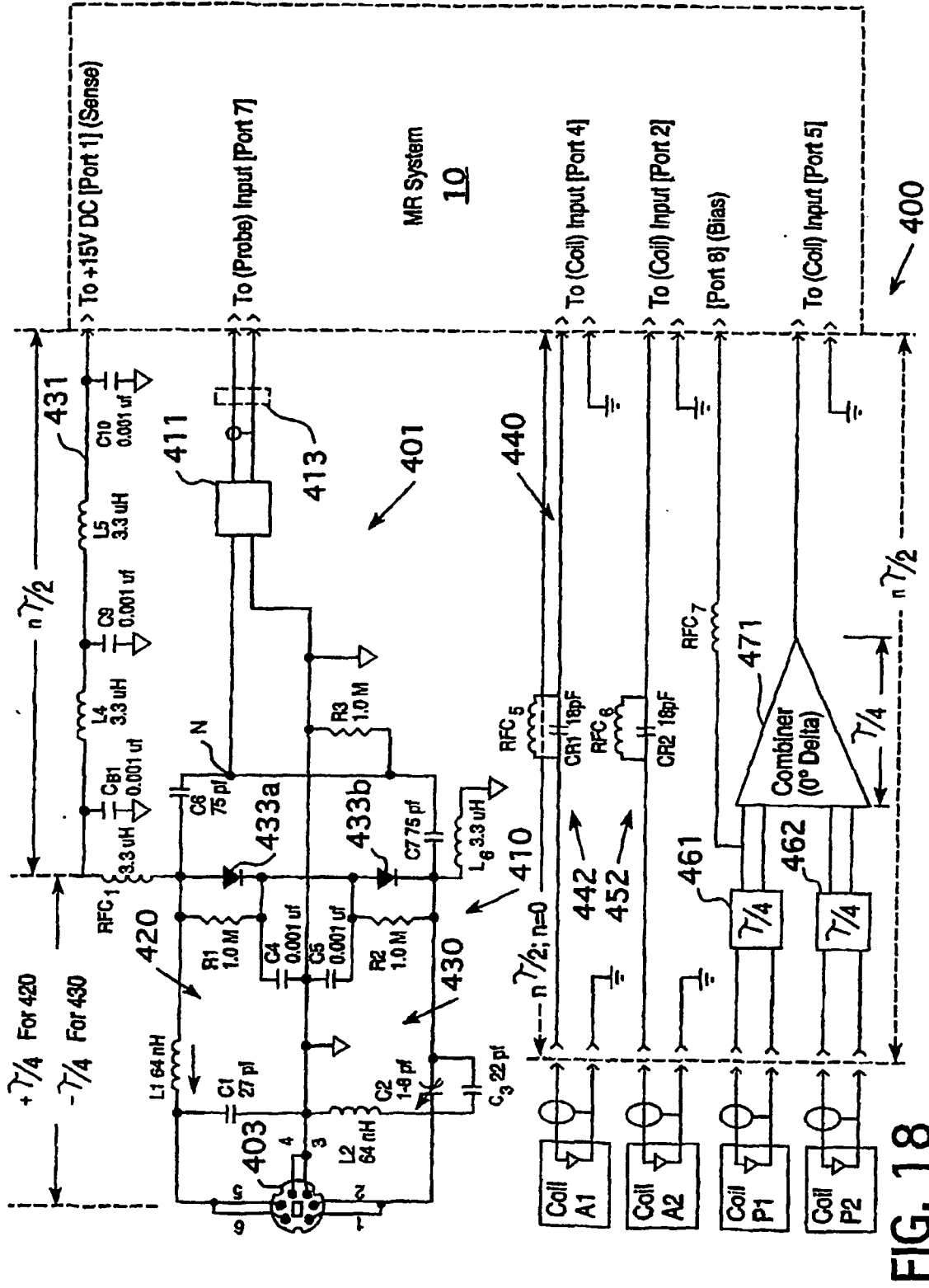


FIG. 18

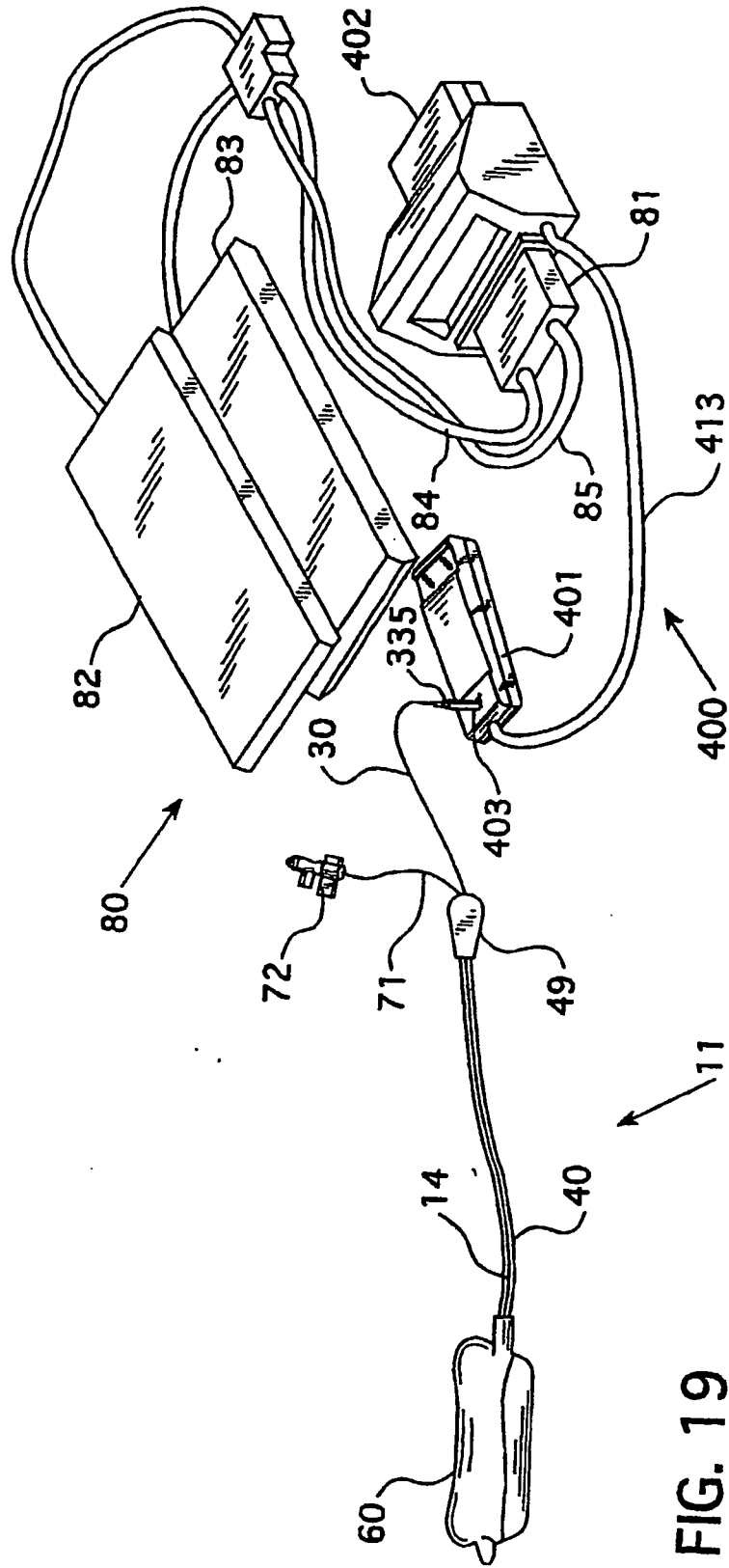


FIG. 19

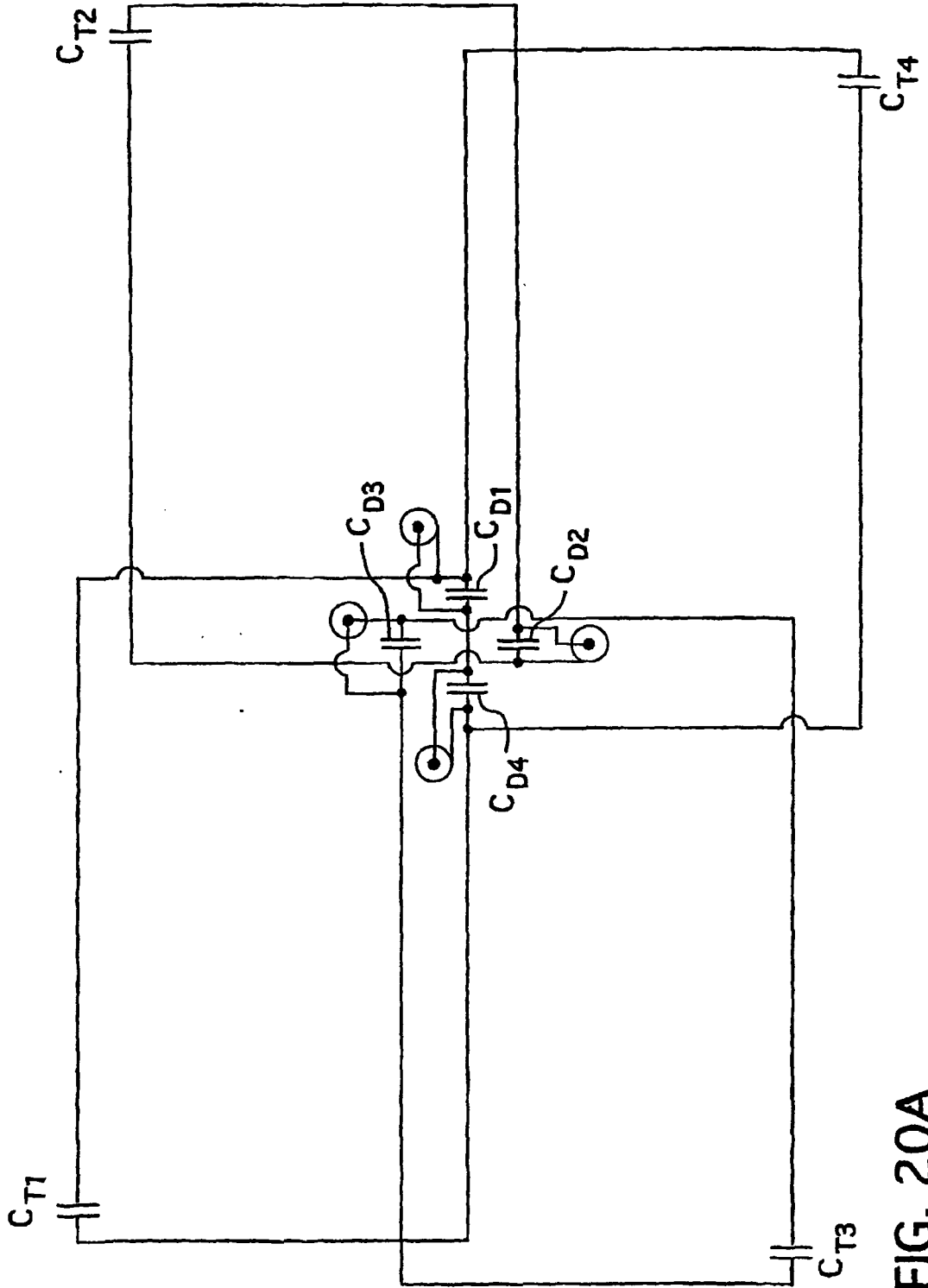


FIG. 20A

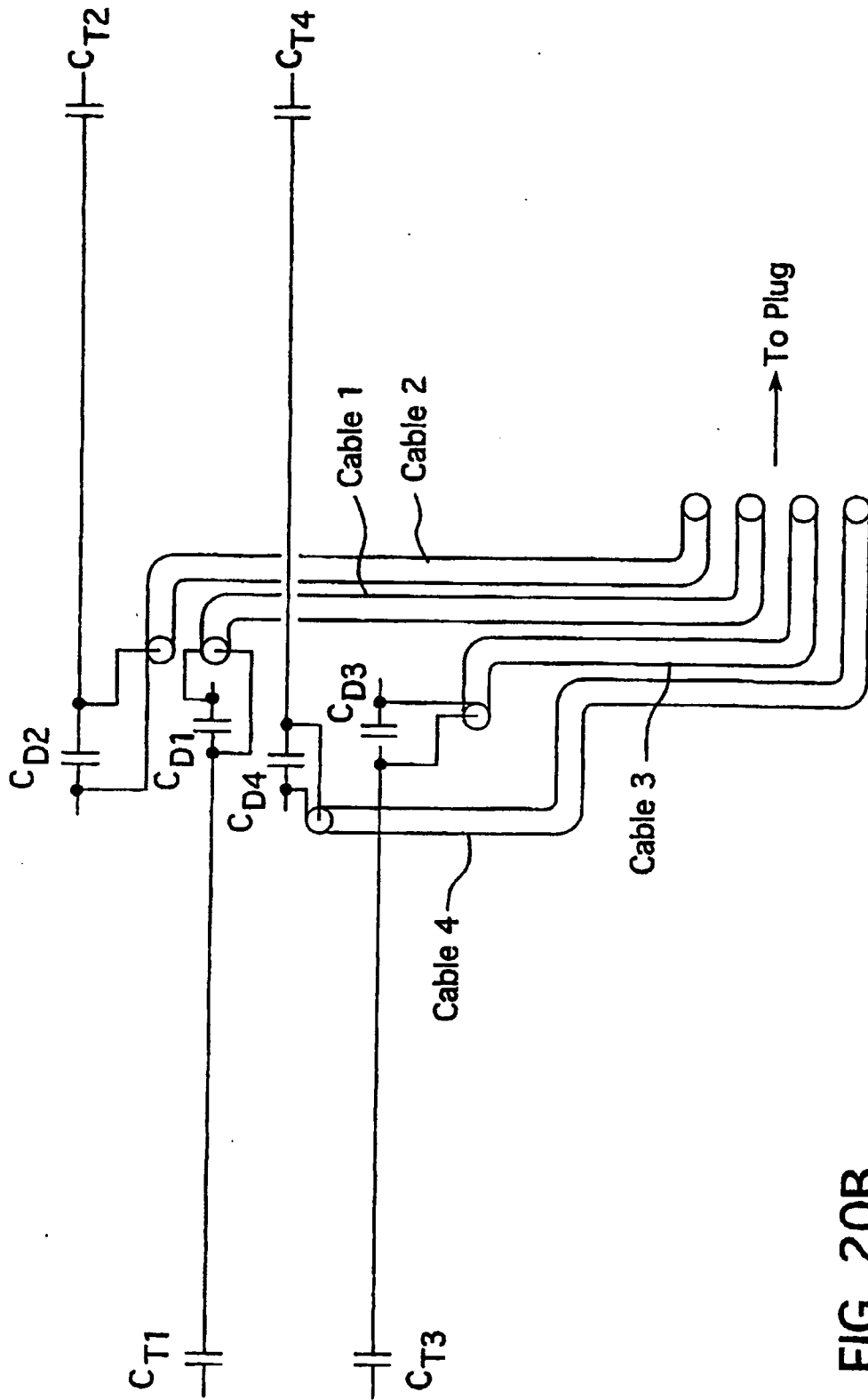


FIG. 20B

REFERENCES CITED IN THE DESCRIPTION

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专利名称(译)	腔内探针及其界面用于使用高场磁共振系统获得腔内结构的图像和光谱		
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申请(专利权)人(译)	MEDRAD INC.		
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外部链接	Espacenet		

摘要(译)

用于连接腔内探针的接口装置 (300) 包括具有接收周期和发送操作周期的磁共振系统 (10) , 所述腔内探针具有线圈环和一对输出电缆 , 所述输出电缆终止于用于连接所述线圈环到所述接口装置 (300) , 所述接口装置 (300) 包括 : (a) 连接器 (303) , 所述腔内探针可通过所述连接器连接到所述连接器 (303) 。 (b) 链接到所述连接器 (303) 的相移网络 (310) , 所述相移网络 (310) 具有分别连接第一和第二所述输出电缆的第一和第二子网 (320,330) 当所述腔内探针和所述接口装置 (300) 的所述连接器 (303) 连接时 , 每个所述输出电缆在其另一端连接在所述线圈环中的驱动电容器上并具有 $SL + n$ 的电气长度 ($\lambda / 4$) 其中 SL 是电抗的补充长度与所述驱动电容器对应的电抗大小相同 , n 是奇整数 , λ 是所述磁共振系统 (10) 的工作频率的波长 ; (c) 一对 PIN 二极管 (33) , 其第一 PIN 二极管连接在所述第一子网 (320) 的输出端 , 第二 PIN 二极管连接在所述第二子网 (330) 的输出端上。这样 , 从所述接口装置 (300) 的所述连接器到与其对应的所述 PIN 二极管的所述第一和所述第二子网 (320,330) 中的每一个的电气长度是 $\lambda / 4$; 其中当所述腔内探针和所述接口装置 (300) 的所述连接器 (303) 连接时 , 所述相移网络 (310) 在所述接收周期期间启用所述线圈环 (i) , 此时所述 PIN 二极管 (33) 被反向偏置为通过所述输出电缆耦合到所述磁共振系统 (10) 的探针输入端口 , 从而允许从每个所述磁共振系统接收磁共振信号输出电缆被构造地组合并路由到所述探针输入端口 , 以及 (ii) 在所述发送周期期间 , 当所述 PIN 二极管 (33) 被正向偏置以通过组合电路从所述磁共振系统 (10) 的发送场去耦合时每个所述输出电缆和与其对应的所述子网 (320,330) 的 $SL + n (\lambda / 2)$ 的长度 (图 16) 。

