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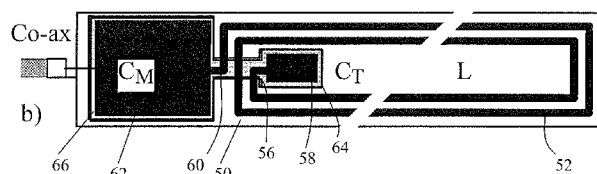
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(54) Title: RADIO FREQUENCY DETECTOR COILS



(57) Abstract: A resonant radiofrequency (RF) detector assembly comprising a substrate, a coil formed on a front surface of the substrate, and two capacitors, each capacitor having a front plate which is formed on the front surface of the substrate and a rear plate formed on a rear surface of the substrate, the two front plates each being electrically connected to a different end of the coil, and the two rear plates being electrically connected to each other.

Radio Frequency Detector Coils

Field of the Invention

The present invention relates to radio frequency (RF) detector coils. It has application, for example, in magnetic resonance (MR) imaging and
5 monitoring systems.

Background to the Invention

Small resonant RF detectors have many applications for *in-vivo* internal magnetic resonance imaging (MRI). Although small coils generally have
10 low Q-factors, this disadvantage is mitigated by the overall increase in signal-to-noise ratio obtained from close coupling to the source. Suitable coil arrangements include single- or multi-turn loops, parallel conductor transmission lines, opposed solenoids and meanders. More compact alternatives for intravascular imaging include the so-called “loopless
15 catheter antenna”, which measures electric rather than magnetic fields. A similar range of coils has been used for the alternative application of catheter tracking.

In each case, the need for matching and tuning has limited widespread
20 clinical application. This problem can be illustrated as follows. Generally, the coil (which has inductance L and resistance R) must be matched to a load R_L at the angular frequency ω_0 at which the coil is to resonate. Figure 1a shows one approach, which uses a first capacitor C_M for matching and a second capacitor C_T for tuning. Standard analysis
25 shows that the parallel arrangement of C_M and R_L may be replaced with a series arrangement of C_M and an equivalent load R_L' , which depends on R_L , C_M and ω_0 , as shown in Figure 1b. Matching requires that $R_L' = R$, which in turn requires that C_M be chosen appropriately. The series sum of C_M and C_T is clearly equivalent to a single capacitor C as shown in Figure
30 1c. Once C_M is fixed, C_T should therefore be chosen so that C is the total capacitance needed for resonance.

Despite advances in modelling methods that allow for 3D coils, skin effects and material losses, it is unfortunately difficult to predict either the resistance R or the inductance L accurately. The coil resistance is inherently frequency-dependent, joints provide further resistive losses, self-capacitance strongly affects resonant frequency, and hand winding introduces variability. Consequently, both C_M and C_T must generally be determined experimentally, using values that are successively updated to improve the degree of matching and the resonant frequency. The restricted set of readily available capacitance values generally forces the use of multiple components for both C_M and C_T . As a result, the final assembly is often bulky, and may have been soldered and re-soldered many times. For large coils, the end product may be acceptable, but the general approach cannot achieve the low cost, small form factor and reproducibility needed for mass deployment of catheter-based probes, especially disposable ones. One solution is to locate the matching and tuning components remotely, using a $\lambda/2$ length of cable. This approach allows a suitable form factor. Matching and tuning can also be carried out using automated varactor-based systems. However, both solutions are generally too complex for low-cost, mass-produced coils.

Micro-fabrication can improve the situation, since it allows repeatable values of R and L to be obtained. Electroplated spiral coils have been formed on rigid substrates such as GaAs, Si and glass. Micro-fabricated Helmholtz coils and gradient coils have been constructed, solenoids have been fabricated on capillaries, planar coils have been integrated with micro-fluidics, and pre-amplifiers incorporated. More recently, attention has turned to flexible plastics such as polyimide and polyether-ether-ketone [Coutrot A.-L., Dufour-Gergam E., Quemper J.-M., Martincic E., Gilles J.-P., Grandchamp J.P., Matlosz M., Sanchez A., Darasse L., Ginefri J.-C. "Copper micromolding process for NMR microinductors

realisation” Sensors and Actuators A99, 49-54 (2002)] and polytetrafluoroethylene [Eroglu S., Gimi B., Roman B., Friedman G., Magin R.L. “NMR spiral surface microcoils: design, fabrication and imaging” Conc. Mag. Res. B 17, 1-10 (2003)], which are much more
5 suitable for *in-vivo* use, and flexible 3D coils have been constructed [Woytasik M., Grandchamp J.-P., Dufour-Gergam E., Martincic E., Gilles J.-P., Megherbi S., Lavally V., Mathet V. “Fabrication of planar and three-dimensional microcoils on flexible substrates” Microsyst. Tech. 12, 973-978 (2006)]. However, matching and tuning have largely been
10 ignored. Some attempts have been made to integrate capacitors, using coplanar conductors [Ellersiek D., Harms S., Casanova F., Blümich B., Mokwa W., Schnakenberg U. “Flexible RF microcoils with integrated capacitor for NMR applications” Proc. MME’05, Göteborg, Sweden, Sept. 4-6, pp 256-259 (2005)], double-layer windings [Woytaskik M.,
15 Ginefri J.-C., Raynaud J.-S., Poirier-Wuinot M., Dufour-Gergam E., Grandchamp J.-P., Girard O., Robert P., Gilles J.-P., Martincic E., Darasse L. “Characterisation of flexible RF microcoils dedicated to local MRI” Microsyst. Tech. 13, 1575-1580 (2007)] or optically-variable MOS structures [Uelzen Th., Fandrey S., Müller J. “Mechanical and electrical
20 properties of electroplated copper for MR-imaging coils” Microsyst. Tech. 12, 343-351 (2006)]. Previously [Ahmad M.M., Syms R.R.A., Young I.R., Mathew B., Casperz W., Taylor-Robinson S.D., Wadsworth C.A., Gedroyc W.M.W. “Catheter-based flexible microcoil RF detectors for internal magnetic resonance imaging” J. Micromech. Microeng.,
25 submitted], it has been demonstrated that high-resolution MRI using catheter mounted microfabricated coils with discrete capacitors, based on the multi-turn rectangular spiral inductor can be produced However, no convincing solution has yet been found to the problem of optimising component values without expensive iteration of planar processing.

Summary of the Invention

The present invention provides a resonant radiofrequency (RF) detector comprising a substrate, an inductor coil formed on a front surface of the substrate, and two capacitors, each capacitor having a front plate formed
5 on the front surface of the substrate and a rear plate formed on the rear surface of the substrate. The two front surface capacitor plates are each connected electrically to a different end of the coil, and the two rear capacitor plates are connected electrically to each other, so that the whole circuit represents a resonant electrical loop containing one inductor and
10 two capacitors.

The resonant circuit may then provide the function of detecting RF signals. One capacitor C_M may then provide the function of matching the electrical impedance seen across its plates at a target resonant frequency
15 to a target value while the other capacitor C_T may provide the function of tuning the resonant frequency of the circuit to a target value.

The coil may comprise one or more full turns, or it may comprise one or more half turns, or part turns, or it may be of any other suitable shape for
20 detecting a RF signal. The front plate of one of the capacitors may be formed within the coil. The front plate of one of the capacitors may be formed outside the coil. The two rear capacitor plates may be formed from a common layer of conductive material. Connections between the two capacitors may also be formed in the same common layer of material.
25 A similar approach may be use to add additional coils and capacitors in series.

The coil may have two sections and each of the front plates may be connected to an end of a respective one of the sections. In some cases the
30 sections each have a respective winding sense, the winding senses being opposite to each other. For example the winding may be arranged in a

figure-of-eight configuration. The sections may each have the same number of turns, or may be otherwise arranged so as to have the same, or substantially the same, inductance. Each of the coil sections may form a respective loop and each of the front plates may be inside a respective one
5 of the loops.

The substrate may be thin, to allow capacitors of a given size to be formed using a small surface area. Using a thin substrate, the inductor and capacitors may be flexible. The whole assembly may therefore be
10 flexible so that it can be wrapped around a catheter.

The present invention further provides a method of producing an RF detector assembly comprising:

- providing a substrate;
- 15 forming a coil on a front surface of the substrate;
- forming two front capacitor plates on the front surface of the substrate;
- forming two rear capacitor plates on a rear surface of the substrate each at least partially aligned with one of the front capacitor plates to form a capacitor
- 20 providing electrical connections between each of the front capacitor plates and a different end of the coil,
- and providing electrical connections between the two rear capacitor plates.

25 The coil and the front capacitor plates may be formed simultaneously, or they may be formed separately in separate steps. The coil and the front capacitor plates may be formed as a common layer of conductive material. The two rear capacitor plates may be formed simultaneously, and may also be formed as a common layer of conductive material. The
30 connection between the two rear capacitor plates may also be formed in this layer. In some embodiments this can allow the whole assembly to be

made using just two steps of patterning a surface conductive layer on a substrate, one layer being provided on the front surface of the substrate and the other layer on the rear surface.

- 5 The present invention further provides a method of producing a resonant RF detector comprising:
- providing a test substrate;
- forming a test coil on a front surface of the test substrate;
- forming two test capacitors each having two plates on opposite surfaces of
- 10 the test substrate;
- providing electrical connection between the capacitors and the coil and each other;
- adjusting the size of at least one of the capacitor plates until the test coil and test capacitors meet a performance criterion;
- 15 determining the size of the test capacitors after the adjusting;
- and subsequently producing the coil assembly so that it has a substrate and coil corresponding in size and shape to the test substrate and coil and two capacitors of the determined sizes.

- 20 In one embodiment the target application is a catheter-based probe for magnetic resonance imaging of the bile duct, but similar approaches would be appropriate for vascular imaging, or for other forms of internal magnetic resonance imaging such as oral, rectal or vaginal imaging that require a small flexible probe.

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The overall aim of some embodiments of the invention is a resonant detector in the form of a flexible sheet that may be wrapped around a catheter and connected to receive electronics, for example via a subminiature co-ax cable.

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In some embodiments a three-stage approach is used. In the first, a RF resonator is formed from a micro-fabricated coil and discrete capacitors. Conventional matching and tuning of this structure allows the values of C_M and C_T to be found. In the second, a fully integrated device is constructed. The same design of coil is used, together with micro-fabricated capacitors whose areas are estimated from the previous experimental values of C_M and C_T . Mechanical trimming of the capacitors following a systematic procedure then allows exact matching and tuning. In the third, a fully integrated device is constructed using an identical micro-fabrication process, but with the now-known capacitor areas. The result is a flexible monolith requiring only connection to a co-axial output, and the method is easily applicable to other coil arrangements.

Preferred embodiments of the present invention will now be described by way of example only with reference to the accompanying drawings.

Brief Description of the Drawings

Figure 1a is a circuit diagram of a detector coil;

Figure 1b is a diagram of an equivalent circuit to that of Figure 1a;

Figure 1c is a diagram of a further equivalent circuit to that of Figure 1b;

Figure 2 is a graph of normalized impedance as a function of normalized coil resistance in the circuit of Figure 1a;

Figure 3 is a graph showing the dependence of capacitance on coil resistance at different field strengths in the circuit of Figure 1a;

Figure 4a is a diagram of a known detector coil assembly;

Figure 4b is a diagram of a detector coil assembly according to an embodiment of the invention;

Figure 5 is a side view of and expanded section through the coil assembly
5 of Figure 2 mounted on a catheter;

Figure 6a is a front view of a set of detector coil assemblies according to the invention;

10 **Figure 6b** shows one of the detector coil assemblies of Figure 6a before and after trimming;

Figure 6c is an enlarged view of the part of the detector coil assembly of Figure 6b;

15

Figure 7 is a graph showing variation of S_{11} for RF detectors with different coil lengths held flat and mounted on a catheter;

Figure 8 is a graph showing variation of S_{11} with frequency for a detector
20 coil assembly according to the invention mounted on a catheter at different stages of matching and tuning, and variation of S_{21} at the final stage;

Figure 9 is a graph showing variation of S_{11} with frequency for a detector
25 assembly according to the invention mounted on a catheter;

Figure 10 is a photograph of a catheter with a detector coil assembly according to an embodiment of the invention mounted on it;

30 **Figure 11** is a photograph of a test phantom;

Figure 12 is a Sagittal ^1H MR image of the test phantom of Figure 11 obtained at 1.5 T, using a fully integrated RF detector with a 40 mm long coil;

- 5 **Figure 13** is a diagram of a detector coil assembly according to a further embodiment of the invention;

Figure 14 is a circuit diagram of the equivalent circuit of the assembly of Figure 13; and

10

Figure 15 illustrates the opposite winding sense of the sections of the coil assembly of Figure 13.

Description of the Preferred Embodiments

- 15 Referring to Figure 1a, according to the standard theory of matching, a lossy inductor with inductance L and resistance R whose impedance at angular frequency ω is $R + j\omega L$ can be tuned to angular resonant frequency ω_0 and matched to a load R_L using two capacitors C_M and C_T .

- 20 The admittance of the combination of R_L and C_M is $Y = 1/R_L + j\omega C_M$. Hence, the corresponding impedance $Z = 1/Y$ can be found as:

$$Z = (1/j\omega C_M) \{1/(1 + 1/j\omega C_M R_L)\} \quad (1)$$

- 25 If $\omega C_M R_L \gg 1$, Equation 1 may be approximated as:

$$Z \approx 1/j\omega C_M + 1/(\omega^2 C_M^2 R_L) \quad (2)$$

- 30 This approximation assumes that $R_L \gg 1/\omega C_M$, i.e. that the load is much greater than the modulus of the impedance of C_M . Equation 2 implies that Z is equivalent to a capacitor C_M in series with a frequency-

dependent load $R_L' = 1/(\omega^2 C_M^2 R_L)$ as shown in Figure 1b. The matching capacitor C_M therefore effectively inserts a scaled version of the load R_L' into the resonant circuit. If C_M is too small, R_L' will be too large, and the overall quality factor of the resonator will be reduced. If C_M is too large, R_L' will be too small and little signal will be detected. Matching involves choosing C_M so that $R_L' = R$ at resonance, which requires:

$$C_M = 1/\{\omega_0 \sqrt{(RR_L)}\} \quad (3)$$

This result implies that $1/\omega_0 C_M = \sqrt{(RR_L)}$, i.e. that the modulus of the impedance of C_M should equal the geometric mean of the two resistors that require matching. C_M must clearly reduce as the frequency rises or as resistive losses increase. We may then rewrite the condition for the earlier approximation as $R_L \gg R$. Since R_L will typically be 50 Ω , and R a few Ohms, Equation 3 will be almost universally valid, independent of the value of L and the operating frequency.

The circuit of Figure 1b is then equivalent to the simple resonator in Figure 1c. Here, C_M and C_T have been combined into a single equivalent capacitor C , whose value $C = 1/(\omega_0^2 L)$ is the total capacitance needed to achieve resonance. Once C_M has been fixed, the tuning capacitor C_T should then be chosen so that $1/C_T + 1/C_M = 1/C$, or:

$$C_T = CC_M/(C_M - C) \quad (4)$$

25

Clearly, C_T must be positive and finite to achieve a meaningful solution. This condition requires $C_M > C$, or $\omega_0 L > \sqrt{(RR_L)}$, so that modulus of the impedance of the inductor must exceed the geometric mean of the two resistors. This condition can normally be satisfied, provided the coil resistance R is low enough. The impedance matching problem may be illustrated graphically as shown in Figure 2. Here we plot the variation of

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the normalised impedance moduli Z/R_L for C_M and C_T , and for an arbitrary assumed inductance L , as a function of the normalised coil resistance R/R_L . Provided R/R_L is small enough and $\omega_0 L$ is large enough, there will be a solution and the relevant approximations will be valid.

5

The operating frequency f_0 for ^1H magnetic resonance (MR) scales linearly with flux at 42.57 MHz/T. In a 1.5 T system (for example), $f_0 = 63.5$ MHz. Figure 3 shows example dependencies of C_M and C_T on the coil resistance R obtained from Equations 3 and 4 at different field strengths, assuming an inductance of 0.35 μH for comparison with later experimental results and a 50 Ω load. At low field strengths, high capacitances are required to reach the relatively low resonant frequency required, and both C_T and C_M vary significantly. As the field strength and resonant frequency rise, C_T reduces considerably and becomes more and more constant. C_M also reduces, but its value still varies significantly, rising rapidly as the coil resistance reduces. In this range, C_M can considerably exceed C_T .

In a first stage of a method of producing detector coils according to an embodiment of the invention, hybrid integrated RF detectors are first produced. Referring to Figure 4a a hybrid integrated detector consists of a substrate layer 10 with a two-turn rectangular spiral winding or coil 12 formed on its surface. The outer end 14 of the winding 12 is connected to the first of four connector pads 16, 18, 20, 22 for surface mount capacitors C_M and C_T . The second and third connector pads 18, 20 are connected together and the fourth connector pad 22 is connected to a further contact 24, on the outside of the winding 12 for connection to one end of an air bridge 26. The inner end 28 of the winding 12 is connected to a further contact 30 for the other end of the air bridge 26, and the air bridge 26 extends over the winding 12 connecting the two air bridge

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contact pads 24, 30. Two capacitors, a matching capacitor C_M and a tuning capacitor C_T , can be connected in series between the first and second 16, 18 and the third and fourth 20, 22 connector pads to form a closed circuit in series with the winding 12 and air bridge 26. Two
5 outputs, for example the inner and outer of a coaxial cable 32, are connected to the first and second connector pads 16, 18 so that the outputs are connected across the matching capacitor C_M .

Referring to Figure 5 the coils of this embodiment were designed for
10 attachment to a catheter 40. The catheter 40 has a tubular body 42 of circular cross section with two lumens 43 extending through it. The coil assembly is arranged so that the long sides 12a of the rectangular coil 12 are parallel to the longitudinal axis of the catheter 40, and the flexible substrate 10 is wrapped round the catheter 40, and a shrink-wrap sleeve
15 44 placed over the coil assembly to hold it in place on the catheter 40. The width of the coil 12, in this case measured from a point between the inner and outer turns on one side of the coil to a point between the inner and outer turns on the opposite side of the coil, is arranged to be equal to half the circumference of the catheter 40. The two sides 12a of the coil 12
20 therefore extend along opposite sides of the catheter 40.

The devices of Figure 4a are fabricated using three lithographic steps. The first defines the majority of the conductors, including the coil 12, the connector pads 16, 18, 20, 22, and the contact pads 24, 30, which are
25 formed simultaneously on the surface of the substrate 10 as a single layer by electroplating inside a photoresist mould. The second step defines a set of holes in an insulating plastic layer formed over the electroplate layer, which allow access to the landing pads and some conductor ends. The third step defines the air bridge 26 that allows connection of the
30 components outside the winding 12 to the inner side of the spiral winding. Coils with this geometry were designed for ^1H MRI at 1.5 T, with the

following parameters: conductor width 200 μm , conductor separation 100 μm , coil length 60 mm, and coil width 4.2 mm. The last value was chosen to place the long conductors on the diameter of an 8 Fr (2.7 mm dia) catheter, when the detector was wrapped around the catheter with its
5 long axis parallel to the catheter.

Coils were fabricated on 25 μm thick polyimide film (Kapton® HN, DuPont, Circleville, OH). This material is mechanically and thermally stable, flexible, pinhole free, resistant to dielectric breakdown, and
10 commercially available in a range of thicknesses [Data Sheet HK-15345: DuPont Kapton (RTM) HN polyimide film" DuPont High Performance Films, Circleville, OH, <http://www.dupont.com>]. To provide a rigid surface for processing, the film was first stretched over a 100 mm diameter silicon wafer and anchored using Kapton (RTM) tape. Seed
15 layers of Ti (30 nm) and Cu (200 nm) metal were then deposited by RF sputtering. A layer of AZ 9260 thick positive photoresist (Microchemicals GmbH, Ulm, Germany) was then deposited by spin coating, and patterned using UV contact lithography to form a mould. 20 μm thick Cu conductor tracks were then formed by electroplating inside this mould, using
20 Technic FB Bright Acid copper plating solution (Lektrachem Ltd., Nuneaton, UK). The mould was stripped, and exposed seed layer was removed by etching. A 2.5 μm thick layer of SU-8 2000 negative epoxy photoresist (Microchem Corp., Newton, MA) was then deposited and patterned to act as an interlayer. Finally, the air bridge was formed, by
25 repeating the steps of seed layer deposition, mould formation, plating, mould removal and seed layer etching. The Kapton® sheet was detached from its carrier, and individual devices were separated using a scalpel. The devices were highly flexible, and could be distorted considerably without conductor detachment.

Electrical performance was measured using an Agilent E5061A network analyser. Matching to $R_L = 50\Omega$ and tuning were carried out with the coil loosely attached with heat shrink tubing to an 8 Fr dual lumen catheter, using discrete 0805 series non-magnetic capacitors (SRT Micro
 5 Céramique, France). The additional components were located just beyond the end of the catheter, and a length of 0.8 mm diameter non-magnetic 50Ω Bluetooth co-axial cable (Axon Cable, Dunfermline, UK) was connected across C_M and passed down one of the catheter lumens. The heat shrink was then tightened to protect the coil, capacitors and cable
 10 joint. Matching and tuning were achieved by minimising the value of the scattering parameter S_{11} at 63.8 MHz using component values of $C_M = 139$ pF, $C_T = 19.5$ pF. Ignoring self-capacitance, these values imply a total capacitance of $C = 17.1$ pF, a coil inductance of $L = 1/(\omega_0^2 C) = 0.36$ μ H and a resistance of $R = 1/(\omega_0^2 C_M^2 R_L) = 6.4$ Ω (and hence a
 15 quality factor of $Q = \omega_0 L/R = 23$). Confirmation of the Q-factor was provided by measurement of the frequency variation of sensitivity for completed resonators.

Referring to Figure 4b, fully integrated resonators were then constructed,
 20 using integrated parallel plate capacitors. These assemblies comprised a substrate 50 of the same material and of the same thickness as that 10 of Figure 4a, and a winding or coil 52 on its front side of the same dimensions and materials as that 12 of Figure 4a. The coil 52 is broken between its inner and outer ends. The inner end 56 of the winding is
 25 connected to a rectangular capacitor plate 58 located within the coil winding 52 and forming part of a tuning capacitor C_T , and the outer end 60 of the winding is connected to a rectangular capacitor plate 62 located outside the coil and forming part of a matching capacitor C_M . Each of the capacitors C_T , C_M also comprises a second, rear plate formed on the rear
 30 of the substrate 50, the two rear plates 64, 66 being connected to each other. The substrate 50 itself therefore is used as the interlayer between

the plates of each capacitor, and as it is formed of thin (12.7 μm) Kapton this allows the plate sizes to be kept small. The two plates 62, 66 of the matching capacitor C_M are connected to an output.

- 5 The coil assembly of Figure 4b is produced using a front side pattern which consists of a spiral, forming the coil 52, linked to the two front capacitor pads 58, 62, while the rear side pattern consists of the pair of similar rear pads 64, 66 linked directly together by a connecting area of conducting material. This arrangement places a larger capacitor C_M
- 10 outside the coil winding, allowing connection of the two plates 62, 66 of that capacitor to an output, which can again be a co-axial output, and a smaller capacitor C_T inside the coil. Because no air bridge is needed to provide connection to the inner end of the spiral, the entire layout may be fabricated using just two lithographic steps to define conductors on either
- 15 side of the substrate. Double-sided lithography and electroplating are required, but front-to-back alignment is not critical since the capacitor plates need only overlap, at least partially, and do not need to be exactly aligned with each other.
- 20 Prototype devices were fabricated using conductor dimensions similar to those above, but a number of coil assemblies in a range of coil lengths were formed on a single substrate as shown in Figure 6a. To allow adjustment of capacitance only by removal of material, capacitors were fabricated using estimated values approximately double those found
- 25 experimentally. Dimensions of 18 mm x 5.5 mm (99 mm²) and 4 mm x 3.5 mm (14 mm²) were used for C_M and C_T ; assuming a relative dielectric constant of 3.5 for Kapton® [“Data Sheet HK-15345: DuPont Kapton® HN polyimide film” DuPont High Performance Films, Circleville, OH, <http://www.dupont.com>], this yielded initial capacitances of 240 pF and
- 30 34 pF respectively. Using processes designed for 100 mm silicon wafers,

14 devices of different length could be fabricated per substrate, as shown in Figure 6a.

Similar patterning and electroplating processes were used to form the
5 conductors on each side. The rear side conductors were formed first, using a smaller Cu thickness (5 μm) and then protected with a layer of photoresist while the front side conductors were formed with the standard Cu thickness (20 μm). The overall thickness was therefore approximately 40 μm . Figure 6a shows a completed substrate as described above, Figure
10 6b shows a pair of completed elements, and before and one after trimming, and Figure 6c an enlarged view in the region of the capacitors, showing part of the coil 12 and the capacitor plates 58, 62 on the front surface of the substrate 10. The resulting devices resonated immediately when placed near an inductive probe.

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Matching and tuning of fully integrated devices was then carried out by mechanically trimming each of the matching and tuning capacitors C_M and C_T , using a systematic process.

20 Completed devices were connected electrically using subminiature co-ax cable, which was soldered across C_M . The initial condition of each device was then assessed, by measuring the frequency variation of the scattering parameter S_{11} with the devices held flat. Figure 7 shows measurements obtained from devices with different coil lengths. In each case, a
25 minimum in S_{11} near a particular frequency implies the existence of a resonance, and the sharpness and depth of the minimum suggests good Q-factor and reasonable impedance matching. The resonant frequency decreases with coil length, and in each case lies below 63.8 MHz. However, the impedance matching reduces as the resonant frequency
30 rises.

The devices were then mounted on a short section of 8 Fr catheter, and the frequency variation of S_{11} was re-measured. This process would be expected to reduce L without changing R , and hence simply increase the resonant frequency. In each case, the matching degraded. Since the effective load is $R_L' = 1/(\omega_0^2 C_M^2 R_L)$, and C_M and R_L are both fixed, the only possible conclusion is that C_M is too large, in line with earlier estimates. The shortest (40 mm) device re-tuned immediately to 63.8 MHz frequency. However, since the longest (60 mm) device only re-tuned to 52 MHz, this device required adjustment.

10

Matching and tuning of the 60 mm device was carried out as shown in Figure 8, which shows frequency variations of S_{11} at different stages. Starting from the initial condition (state 1), C_M was first trimmed with a scalpel. After several adjustments, state 2 was reached. Here, the depth of the minimum in S_{11} suggests excellent matching, but at the wrong resonant frequency (which has only risen slightly, to 53 MHz). However, because impedance matching occurs at a particular value of $\omega_0 C_M$, the capacitance C_M needed to match at a one angular frequency (ω_0) can be used to estimate the value C_M' needed at another (ω_0'), as $C_M'/C_M = \omega_0/\omega_0'$. It was therefore simple to estimate the optimum area of matching capacitor required at 63.8 MHz, and C_M was trimmed appropriately to state 3. Here the resonant frequency has again risen slightly (to 54 MHz) and the impedance matching appears to have degraded. The tuning capacitor C_T was now trimmed through states 4 and 5, gradually increasing the resonant frequency and improving the matching, to the final state 6. At this point, the accuracy of the tuning and matching are both excellent. The Q-factor of tuned and matched device was again estimated as 22, by measuring the frequency variation of S_{21} using an inductive probe as shown in Figure 8.

30

The final areas of C_M and C_T were 52 mm² and 9.3 mm², respectively, corresponding to estimated capacitances of 127 pF and 23 pF (close to the target values). These results show that a method for identifying the initial state of an unknown device exists, and (provided the two capacitances are both too large), there is a convergent algorithm to match and tune by reduction of overlap areas. However, if necessary adjustment can still be carried out when capacitors are too small, simply by extending overlap areas with silver-loaded epoxy.

It can be a problem that mechanical trimming can short one of the capacitors, due to small slivers of metal tracking between the closely spaced conductors. This problem can be eliminated, by replacing mechanical trimming of the entire structure with laser trimming of just one, or both, of the conductors, so that an insulating dielectric layer remains in place throughout. In addition, excess solder flow during cable attachment can spoil the flexibility of the metal layers. This problem can be addressed using additional patterned surface layers to limit the solderable area.

Once the final area is known, a mask redesign may be carried out to fabricate further devices whose initial state is even closer to the desired final state. Figure 9 shows the frequency variation of S_{11} of a device fabricated with the capacitor areas determined from the capacitor trimming and testing described above. This device is inherently tuned and matched, and requires only a soldered connection to an output to operate. Figure 10 is a photograph of a completed catheter-based RF micro-coil detector.

¹H magnetic resonance imaging experiments were carried out on phantoms, using resonators as described above with reference to Figures

6a to c to demonstrate the imaging capability of these integrated resonators.

Imaging was performed using a 1.5 T GE HD Signa Excite scanner. The
5 system body coil was used for transmission and a 40 mm long catheter
mounted micro-coil as shown in Figure 5 was connected to the auxiliary
coil input for signal reception. No measures (such as diode-switched
detuning) were taken to avoid damage by the transmit pulse, since
previous experience suggested this to be unnecessary. The micro-coil was
10 initially placed on a large spherical phantom containing a dilute solution
of NiCl_2 and CuSO_4 . The micro-coil was located at the isocentre with the
long conductors lying in the coronal plane, and autotuned using a fast
recovery fast spin echo (FRFSE) sequence.

15 High-resolution imaging was then demonstrated using a phantom
consisting of an M4 nylon nut and cheese-head bolt (tooth pitch 0.7 mm),
which was placed in solution in a small glass cuvette as shown in Figure
11. The cuvette was placed on top of the coil, which remained on top of
the spherical phantom. Imaging was carried out using a relaxation
20 recovery time (TR) of 33 ms, an echo time (TE) of 15 ms and a flip angle
of 10° . The images were acquired using a T_2 -weighted FRFSE sequence in
28 slices of 1.2 mm thickness, with 256 x 224 pixels per slice in an 80
mm x 40 mm field of view. With 6 excitations to improve SNR the total
acquisition time was 11 min 53 sec.

25

Figure 12 is a typical sagittal slice, which shows the cuvette containing
the nut and bolt at the top of the figure and the spherical phantom at the
bottom. The slotted head of bolt is clearly visible, and in the original
images the teeth may also be seen. These results are comparable to those
30 previously obtained in with partially integrated coil assemblies as shown
in Figure 4a [Ahmad M.M., Syms R.R.A., Young I.R., Mathew B.,

Casperz W., Taylor-Robinson S.D., Wadsworth C.A., Gedroyc W.M.W.
“Catheter-based flexible microcoil RF detectors for internal magnetic
resonance imaging” J. Micromech. Microeng., submitted], and
demonstrate that the imaging performance of fully integrated coils can be
5 as good as that of coils with surface mount capacitors.

In addition to providing a means of realising spiral coils with integrated
tuning and matching capacitors, as described in the embodiments above,
other embodiments of the invention make use of its ability to allow more
10 complex coil arrangements, which can have advantages in certain
applications.

Referring to Figure 13 in a further embodiment, a two-turn, figure-of-
eight coil with tuning and matching capacitors is provided. The circuit is
15 again constructed from a thin substrate layer 101 with the coil windings
102 formed on its front surface, from a first track 102a and a second
track 102b. The first track 102a extends from the matching capacitor
front plate 103 and spirals inwards through one and a half turns, to form
the first loop 102a of the winding, and ends with a first interconnection
20 capacitor plate 105 inside the first loop. It therefore forms all of the first
loop except one half turn. The second track 102b extends from the tuning
capacitor front plate 104 along the substrate adjacent to the first track to
form a half turn of the first loop 102a, and then spirals inwards through
two complete turns to form the whole of the second loop 102b of the
25 figure-of-eight, ending with a second interconnection capacitor plate 106
inside the second loop. An interconnection bridge 108 is formed on the
back surface of the substrate 101 and comprises two plates 108a, 108b
each of which is opposite one of the interconnection capacitor plates 105,
106, coupled together by a short track 108c. A further plate 107 on the
30 back surface of the substrate 101 forms the back plates of the matching

and tuning capacitors. Both the tuning capacitor and the matching capacitor have their front plate 103, 104 outside the coil 102.

The resulting winding on the front face of the substrate has two loops
 5 each having two full turns, the winding being broken in the outer turn at one end of the first loop, to allow connection to two plates 103 and 104 and also broken at its centre where the resulting gap, between the two capacitor plates 105 and 106 which form ends of the coil on the front face, is bridged by the interconnecting bridge 108.

10

The arrangement of Figure 13 has the equivalent circuit shown in Figure 14. The coil 102 provides an inductance L , divided into two sections L_1 and L_2 . The plates 103 and 107 together with the substrate 101 provide a matching capacitor C_M , while the plates 104 and 107 together with the
 15 substrate 101 provide a first tuning capacitor C_{T1} . The plates 105 and 108a together with the substrate 101 provide a second tuning capacitor C_{T2} , while the plates 106 and 108b together with the substrate provide a third capacitor C_{T3} . The angular resonant frequency of the circuit is then:

$$20 \quad \omega_0 = 1/(LC)^{1/2} \quad (5)$$

Where L is the total inductance and C is the total capacitance, and L and C are given by:

$$25 \quad L = L_1 + L_2 \quad (6)$$

$$1/C = 1/C_M + 1/C_{T1} + 1/C_{T2} + 1/C_{T3} \quad (7)$$

It will be appreciated that the circuit can operate as a resonant detector for RF signals, and that matching and tuning may be carried out as
 30 described above for the embodiment of Figure 4b. It will also be appreciated that the function of the capacitors C_{T2} and C_{T3} is to allow a

figure-of-eight coil arrangement to be realised from two patterned conducting layers formed on either side of an insulating substrate, without the need for an air bridge or a via connection. It will also be appreciated that the winding of the coil is divided into two halves, with an opposite winding sense in each half, as shown in Figure 1c. Each half, or loop, has the same number of turns.

It will be appreciated that if the inductors L_1 and L_2 are identical, a uniform time-varying external magnetic flux B_1 acting perpendicular to the coil will induce equal and opposite emf in each half winding, which will therefore cancel to yield zero net emf and zero current. Consequently, the coil will have low sensitivity to a spatially uniform RF magnetic field, such as the field generated by the body coil of a MRI scanner during excitation. As a result, directly induced voltages and local modification of the excitation pattern may be minimised. However, the coil can still have sensitivity to the locally generated RF fields that arise during signal reception.

This feature provides an inherent passive de-coupling between the transmitter and receiver of the MRI system, which can avoid the need for other methods of de-coupling such as diode-switched de-tuning. Consequently, the arrangement can provide a de-coupled coil in thin-film form, that can be fabricated entirely by patterning of conductor layers without the need for additional semiconductor components.

Finally, it will also be appreciated that the use of paired capacitors to allow a figure-of-eight coil winding without the use of an airbridge can be extended to provide windings that are further subdivided into additional sections, providing a multi-section coil whose winding alternates in sense between adjacent sections or loops. If there are an even number of sections, then the induced emfs in the sections can be balanced by making

all of the loops the same size and with the same number of turns. In other cases the sections can be of different sizes, and there may be different numbers of sections with the two winding senses, but with the correct selection of coil size and shape, the emfs can still be balanced.

Claims

1. A resonant radiofrequency (RF) detector assembly comprising a substrate, a coil formed on a front surface of the substrate, and two capacitors, each capacitor having a front plate which is formed on the front surface of the substrate and a rear plate formed on a rear surface of the substrate, the two front plates each being electrically connected to a different end of the coil, and the two rear plates being electrically connected to each other.
2. An assembly according to claim 1 wherein the front plate of one of the capacitors is formed within the coil.
3. An assembly according to claim 1 or claim 2 wherein the front plate of one of the capacitors is formed outside the coil.
4. An assembly according to any foregoing claim wherein the two front capacitor plates and the inductor are formed from a common layer of conductive material.
5. An assembly according to any foregoing claim wherein the two rear capacitor plates are formed from a common layer of conductive material.
6. An assembly according to any foregoing claim wherein the substrate, coil and capacitors are flexible.
7. An assembly according to any foregoing claim wherein the coil has two sections and each of the front plates is connected to a respective one of the sections.

8. An assembly according to claim 7 wherein the sections each have a respective winding sense, the winding senses being opposite to each other.
- 5 9. An assembly according to claim 8 wherein the sections each have the same number of turns.
- 10 10. An assembly according to any of claims 7 to 9 wherein each of the coil sections forms a respective loop and each of the front plates is inside a respective one of the loops.
- 15 11. A method of producing a resonant RF detector comprising: providing a substrate; forming a coil on a front surface of the substrate; forming two front capacitor plates on the front surface of the substrate; forming two rear capacitor plates on a rear surface of the substrate each at least partially aligned with one of the front capacitor plates to form a capacitor, and providing electrical connection between the front capacitor plates and the coil and between the two rear capacitor plates.
- 20 12. A method according to claim 11 wherein the coil and the front capacitor plates are formed simultaneously on the front surface of the substrate as a common layer of conductive material.
- 25 13. A method according to claim 7 or claim 8 wherein the rear capacitor plates are formed simultaneously on the rear surface of the substrate as a common layer of conductive material.
- 30 14. A method of producing a resonant RF detector comprising: providing a test substrate; forming a test coil on a front surface of the test substrate; forming two test capacitors each having two plates on opposite surfaces of the test substrate; and providing electrical connection between

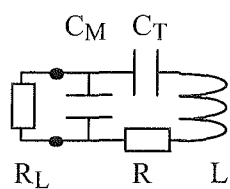
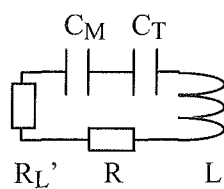
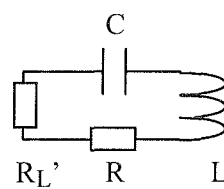
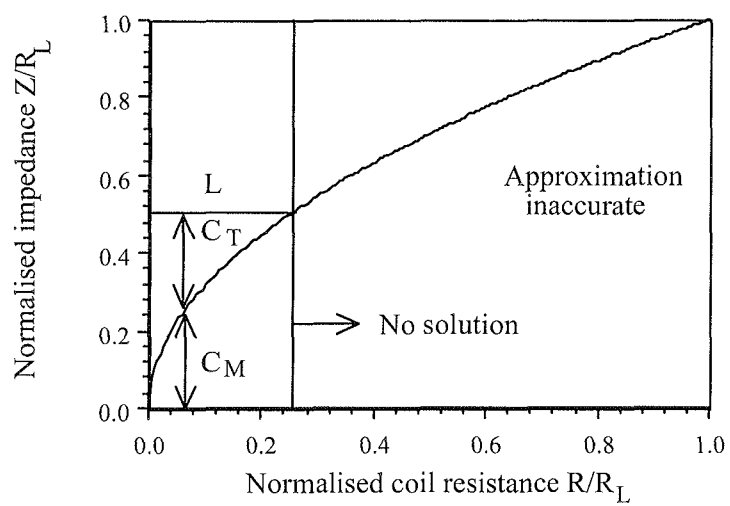
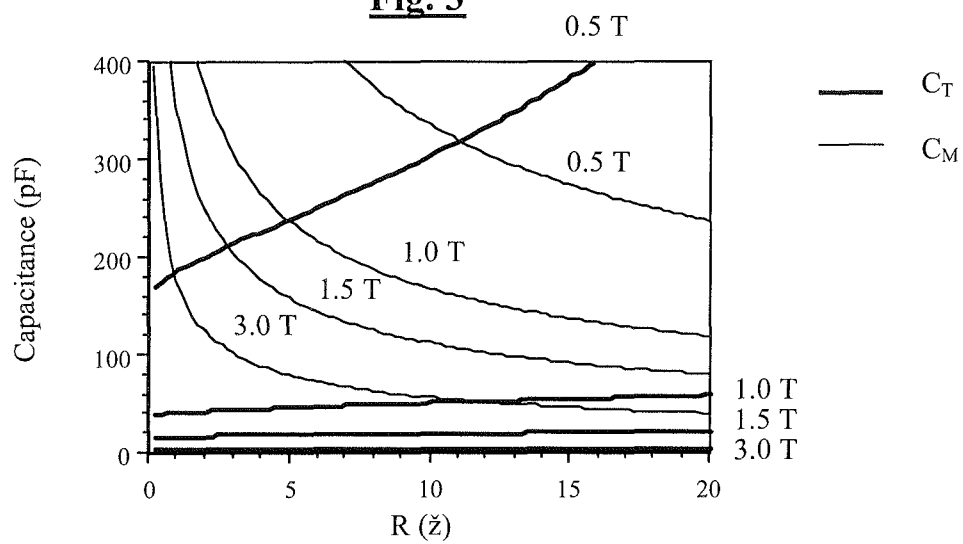
the capacitors and the coil and between each other; adjusting the size of at least one of the capacitor plates until the test coil and test capacitors meet a performance criterion; determining the size of the test capacitors after the adjusting; and subsequently producing the coil assembly so that it has
5 a substrate and coil corresponding in size and shape to the test substrate and coil and two capacitors of the determined sizes.

15. A method according to claim 14 wherein the size of at least one plate of each of the capacitors is adjusted.

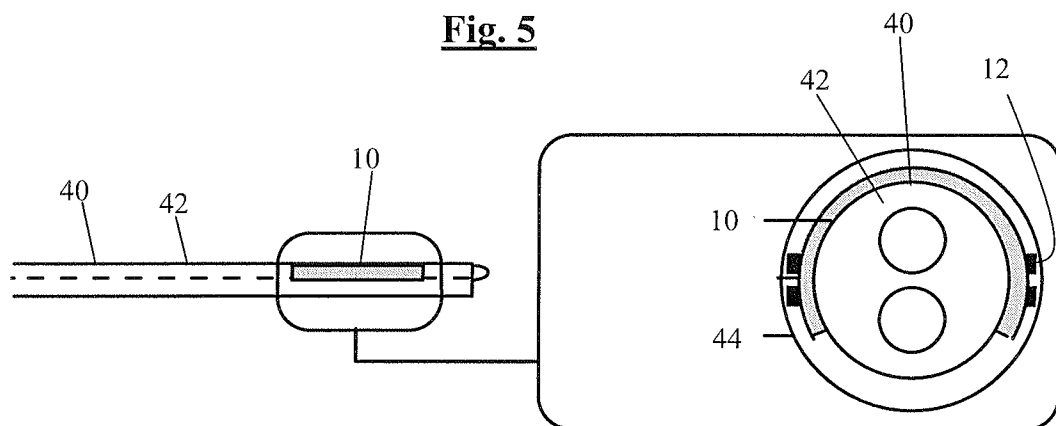
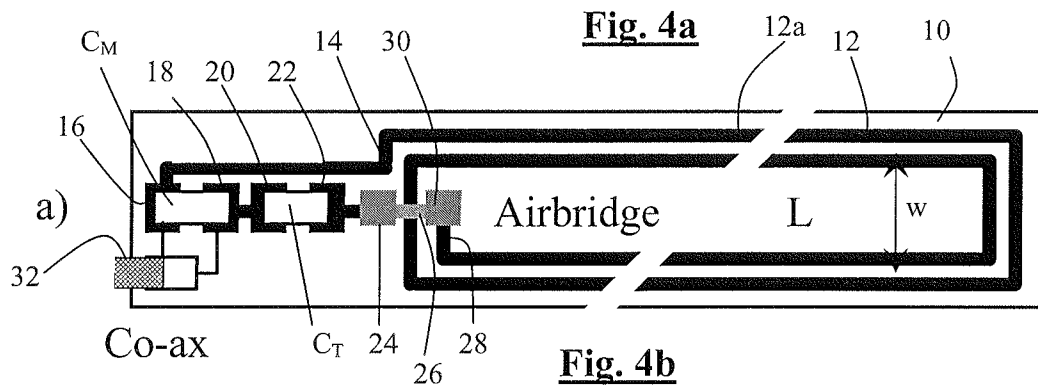
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16. A method according to claim 14 or claim 15 wherein the performance criterion comprises at least one of: a degree of matching the impedance seen across one capacitor at a predetermined frequency to a predetermined electrical load, and a degree of tuning the resonant
15 frequency of the circuit to a predetermined value.

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**Fig. 1a****Fig. 1b****Fig. 1c****Fig. 2****Fig. 3**

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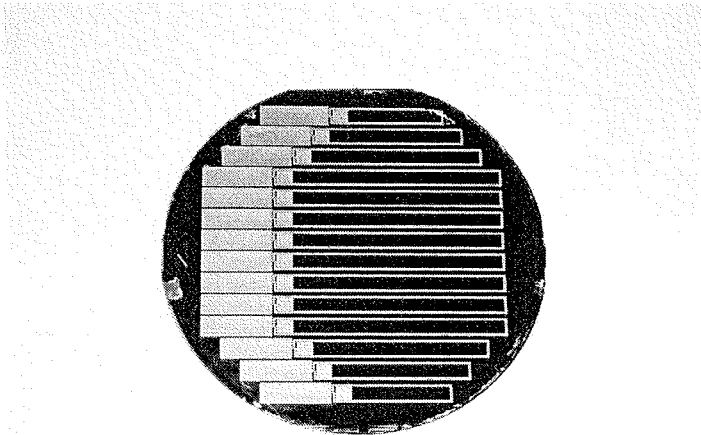


Fig. 6a

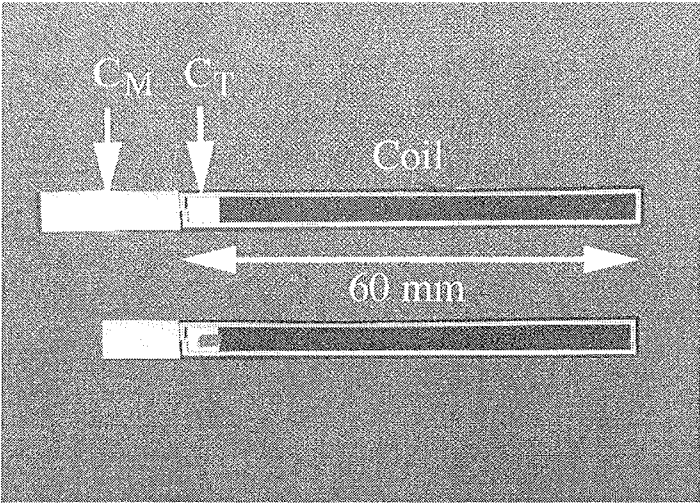


Fig. 6b

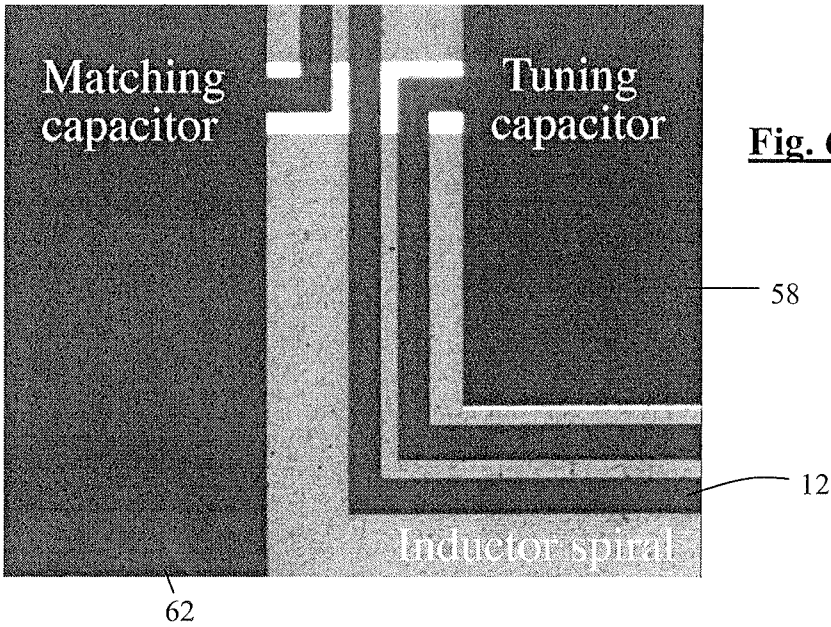


Fig. 6c

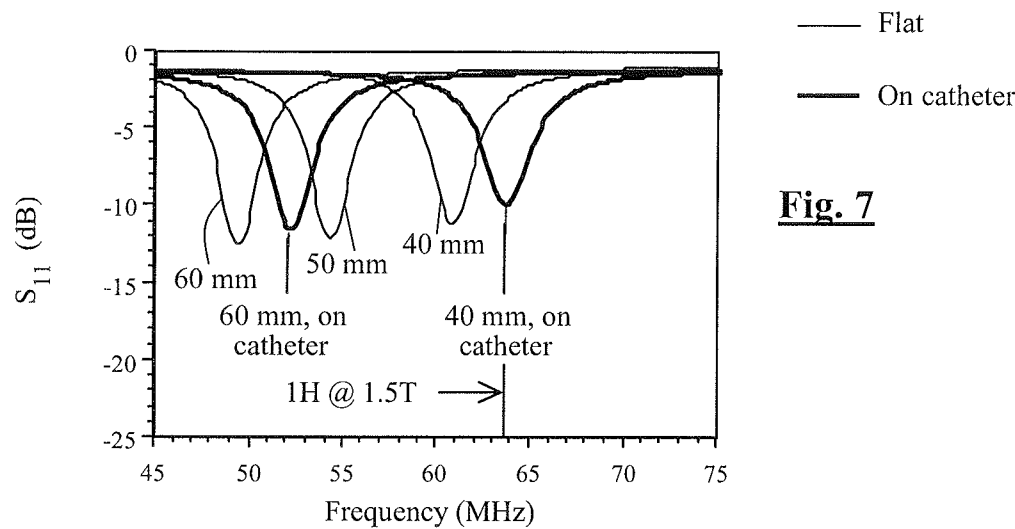


Fig. 7

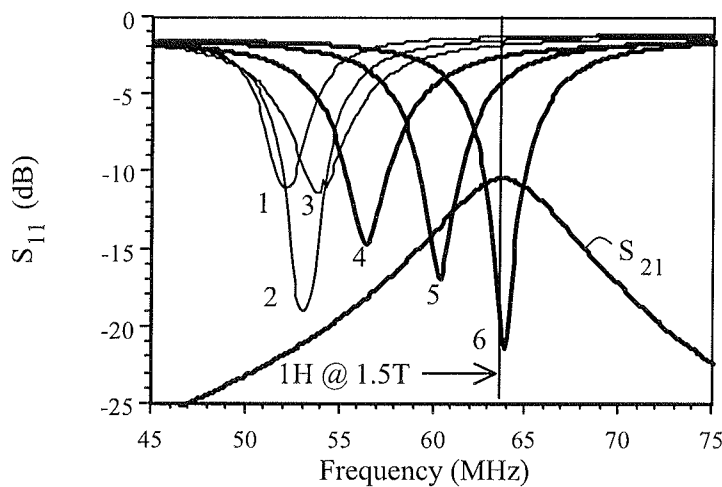


Fig. 8

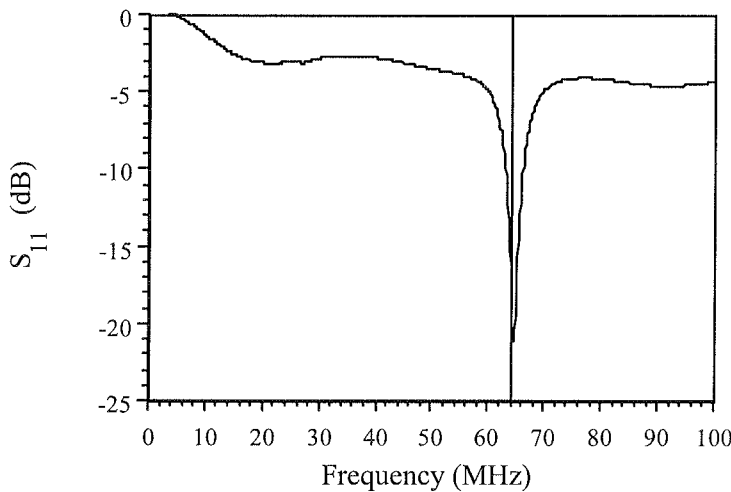


Fig. 9

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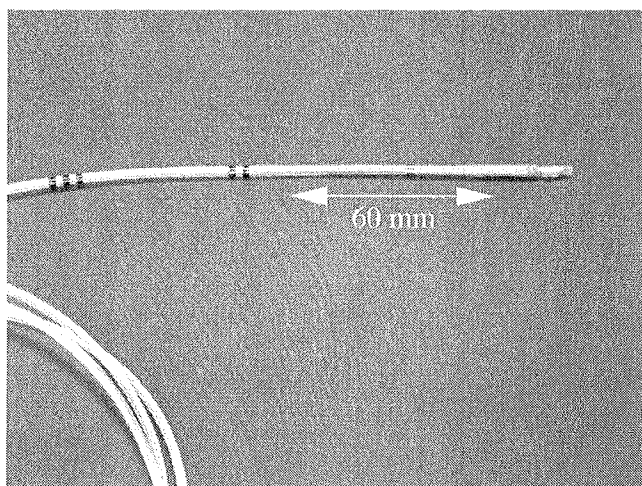


Fig. 10

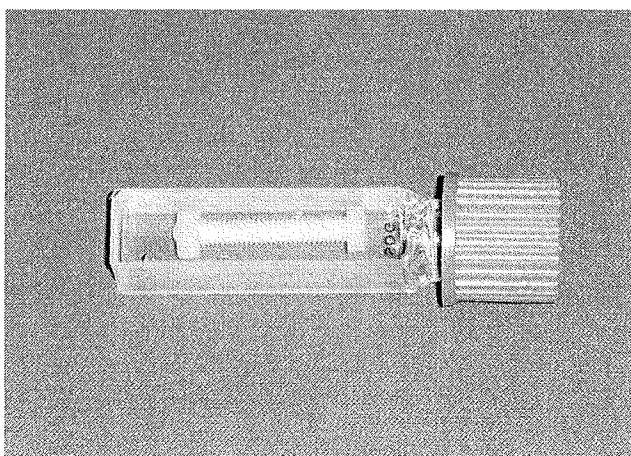


Fig. 11

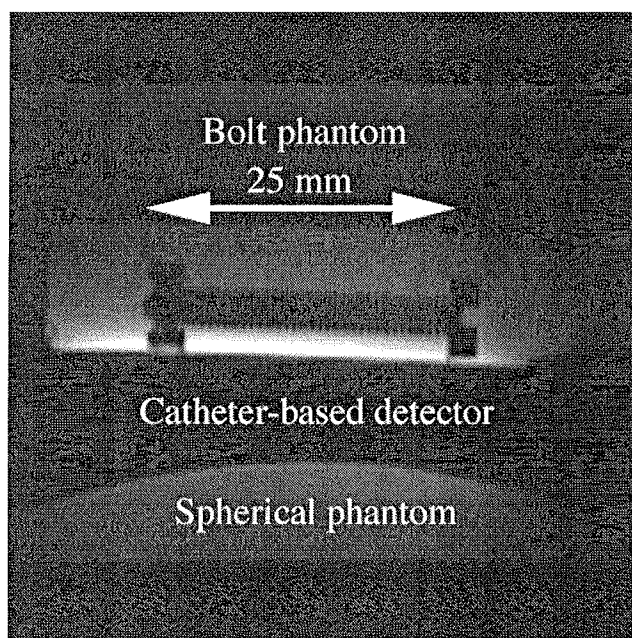


Fig. 12

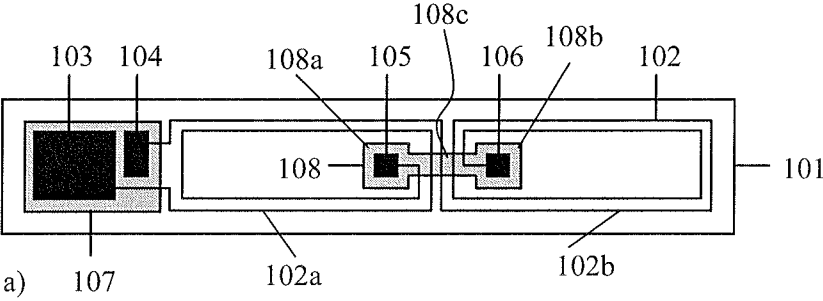


Fig. 13

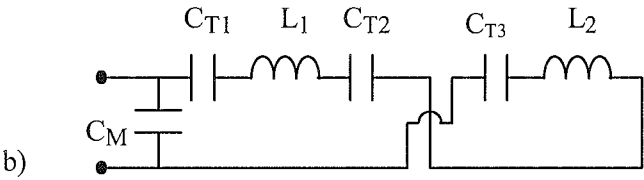


Fig. 14

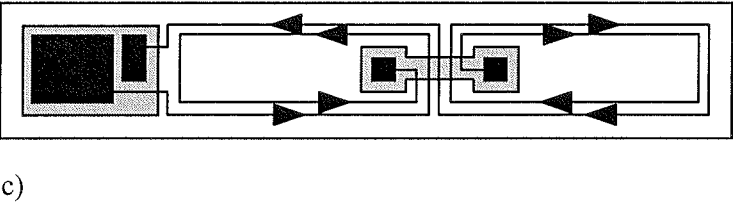


Fig. 15