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Signal Sensitivity Enhancement of High-Spatial-Resolution MR Imaging with a Concatenated Cylindrical Parametric RF-Resonator

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Abstract

Phased array MRI coils can increase sensitivity of superficial tissues owing to their proximity to the detection region. Deep-lying tissues, on the other hand, do not benefit to the same degree. Here we investigate the use of a localized cylindrically symmetric quadruple frequency resonator concatenated with a double frequency resonator to increase the longitudinal field-of-view (FOV) without compromising the spatial-resolution and detection sensitivity. These concatenated array coils work on the principle of a parametric amplification to provide wireless amplification of the locally detected NMR signal prior to inductively coupling the coil to an external pick-up loop with connection to the system receiver. When both the detectors are activated together, the effective range of both overlay to create a larger FOV enabling better identification of detectable regions. Furthermore, the *in-vivo* test of the concatenated detector provides a worst-case 5-fold SNR gain in regions separated from the cylindrical surface larger than its own diameter. This proposed approach of concatenated detector realization can be individually activated and manipulated to enlarge the sensitivity-enhanced region without sacrificing their individual performance. Compared to double frequency detectors, quadruple frequency detectors offer more flexibility in the choice of detector dimension, enabling multi-element concatenation over an extended FOV.

Keywords

Concatenation; Extended Field-of-View (FOV); Parametric Amplification; Sensitivity Enhancement; Wireless

I. Introduction

MAGNETIC resonance imaging (MRI) is a versatile medical imaging technology for disease diagnosis. The quality of magnetic resonance (MR) images plays a crucial role in the in-depth study of physiologies and small pathologies that need to be diagnosed. This appears to be challenging when imaging targeted deep lying tissue structures due to their lower detection sensitivity [1], [2]. Our work, hence, focuses on enhancing the detection sensitivity of deep-lying structures during MR imaging, with a flexibility to extend the sensitively-detected region. Our approach has benefited from the fact that the localized coil placed in the vicinity of the regions-of-interest (ROIs) has better sensitivity with a significant increase in signal-to-noise ratio (SNR) [3]. But instead of using wire connectors [4], [5] for signal transmission, this method uses the novel Wireless Amplified Nuclear MR Detector (WAND) [6]-[8], thus removing the constraints imposed by physical links used for signal propagation. The WAND has an integrated amplifier that exploits the parametric amplification [9]-[11] process to transfer energy from the high energy pumping signal to the low energy MR-input signal [12], [13], thus greatly enhancing the local sensitivity by amplifying weak MR signals before wirelessly transmitting them to an external receiver coil [14], [15]. This procedure does not require an internal power source because the weak initial MR signal can exchange energy with the strong pumping signal that is provided wirelessly, thus significantly improving the signal transmission efficiency over a longer-distance separation and the operation flexibility [16].

Previous designs of WANDs are mostly double frequency resonators [17]-[19] with cylindrical symmetry that used the high frequency resonance mode to receive the pumping signal and the low frequency resonance mode to receive the MR input-signal. As required for efficient parametric signal amplification, the pumping frequency denoted as ω_3 had to be approximately twice the MR detection frequency ω_1 , i.e., $\sim 2\omega_1$, which required a small ratio (~ 2.8) between length and diameter, restricting the achievable longitudinal field-of-view (FOV). This limitation mandates that cylindrically symmetric double frequency resonators to be used as stand-alone detectors rather than feasibly concatenating them for extended FOV, because whenever they are concatenated and coupled with the same pumping signal, it is virtually impossible to wirelessly optimize their individual performance by a single control knob [17], [18], [20]. In this work, we propose a cylindrically symmetric quadruple frequency resonator to overcome the limitation of smaller length-to-diameter ratio mandated by double frequency resonators. Since the quadruple frequency resonator design can be activated at a pumping frequency different from $2\omega_1$, we can concatenate the double frequency resonator with a quadruple frequency resonator to further enlarge the longitudinal FOV. This is possible because when the quadruple frequency resonator's highest resonance frequency ω_3 is approximately equal to the sum of its two lower resonance frequencies $\omega_1 + \omega_2$, efficient amplification can occur, where ω_2 is called the 'idler' frequency or the difference frequency given by $\omega_2 = \omega_3 - \omega_1$. Even though parametric amplification requires only three resonance signals, we will introduce four frequency modes into the cylindrically symmetric quadruple resonator to enable easy adjustment of its high-frequency butterfly mode (ω_3) by the center capacitor whose capacitance change will not affect other resonance modes. Because different quadruple frequency resonators can be engineered to be responsive

to different pumping frequencies, it is possible to concatenate multiple resonators together and use a specific pumping frequency to individually manipulate each component. The goal of this work is to demonstrate this possibility.

II. Methods

A. Working Principle of the Parametric Amplifier (Paramp)

Wireless Amplified NMR detectors can wirelessly transmit MR signals, aided by an on-board amplifier that is also wirelessly powered [21], [15]. In contrast to transistor-based low-noise amplifiers that require a wired direct current (DC) power source [22], parametric amplifiers can be powered wirelessly with an externally applied radio frequency (RF) field such that the internally detected signal is amplified *in situ* before it is wirelessly transmitted by an external detector. This so called ‘pumping signal’ amplifies the weak MR signal at the MR frequency ω_1 as well as the signals at the difference frequency ‘idler’ between the pumping signal and the MR input signal $\omega_2 = \omega_3 - \omega_1$ [23], [24]. During this process, energy is supplied to the circuit through capacitance variation of a PN-junction diode that amplifies both the signal (ω_1) and the idler (ω_2) frequencies. The simultaneous reception and subsequent amplification of weak NMR signals at frequency ω_1 with sufficient gain from the WAND is possible when the following two conditions are satisfied. In the first condition, it requires that the relative frequency offset at the signal frequency ω_1 should be equal to the relative frequency offset at the idler frequency ω_2 . This is called the frequency matching condition and is given by the relation (1) [16]:

$$\frac{X_1}{R_1} = \frac{2Q_1(\omega_1 - \omega_{10})}{\omega_{10}} \approx \frac{X_2}{R_2} = \frac{2Q_2(\omega_2 - \omega_{20})}{\omega_{20}} \equiv \kappa, \quad (1)$$

where ω_{10} , and ω_{20} are the resonance frequencies of the local resonator and Q_1 and Q_2 are the respective quality factors or Q -factors at these frequencies. ω_1 is the Larmor frequency of the MR scanner and ω_2 is the idler frequency, $\omega_2 = \omega_3 - \omega_1$. Moreover, R_1 and X_1 are the resistance and the reactance of the circuit loop that contains the varactor diode at frequency ω_{10} with effective impedance $Z_1 = R_1 + jX_1$. Similarly, R_2 and X_2 are the resistance and reactance of the circuit loop at frequency ω_2 with its effective impedance $Z_2 = R_2 + jX_2$. Another condition requires that the pumping energy should be sufficient in higher resonance frequency mode ω_3 so that it yields sufficient negative resistance at the lower resonance frequency ω_1 , which then cancels part of the resistive loss, increasing the resonator’s Q -factor and thus its signal intensity. Circuit has always some kind of losses due to positive resistance. But if circuit has amplification by some means, we can consider the regeneration energy as negative resistance. As a result, a weak MR signal at ω_1 mixes with a strong pumping signal at ω_3 provided wirelessly to amplify and create the output idler frequency ω_2 , which further can mix back with the pumping signal to produce a second amplified output at ω_1 [25]-[27]. This amplified signal is then detected by a standard external coil [28]. To make the paramp useful over a large enough bandwidth, we need to slightly reduce the negative resistance by decreasing the pumping power beneath the resonator’s oscillation threshold.

B. Resonance Frequency Modes

To achieve efficient parametric amplification from the single element double frequency resonators, its longitudinal mode which is sensitive to a strong pumping signal at ω_3 should be double of its transverse resonance mode to receive weak MR detection frequency at ω_1 , as idler frequency ω_2 approximates the Larmor frequency ω_1 and thus share the same circuit mode in the resonator. A previous version of a parametric amplifier is implemented as a cylindrical double frequency resonator with two split end rings and two vertical legs [18]. The two vertical legs create the lower-frequency mode (also called the horizontal dipole mode), which is sensitive to spin precession in the transverse plane and the two end-rings whose gaps are bridged by two identical diodes create the higher-frequency mode (also called the ring mode), which is sensitive to a pumping field applied longitudinally. The cathode of one varactor diode is connected to the anode of other so that the capacitance of both diodes is modulated in the same direction for efficient signal amplification when the longitudinal pumping field is applied [29]. The continuous center ring is connected to the virtual ground of both legs, so it will have little effect on the dipole-mode resonance. Instead, it is included to stabilize the circuit by effectively neutralizing the accumulated charge across the varactor diodes at the end of transmit pulse. The resonator's low-frequency horizontal mode is centered around the Larmor frequency ω_1 and its high-resonance ring mode is centered around twice the Larmor frequency to receive the pumping signal at a frequency ω_3 slightly higher than $2\omega_1$. The frequency offset between ω_3 and $2\omega_1$ should be at least the imaging bandwidth, so that the "idler signals" created at the difference frequency, $\omega_2 = \omega_3 - \omega_1$, can be filtered out to eliminate destructive interference with MR signals at ω_1 [16], [18], [20], [30]. Because the 2-to-1 frequency ratio between the higher and lower frequency modes also mandates a relatively small dimension ratio (~ 2.8) between length and diameter, a small-element double frequency resonator itself is not sufficient to observe an extended longitudinal field-of-view.

To overcome this limitation, we have created a quadruple frequency resonator with larger length-to-diameter ratio. The circuit design is similar to the double frequency resonator with two split end rings and two vertical legs. However, the center continuous ring is split by a chip capacitor in the quadruple frequency resonator (Fig.1) to independently create a third resonance mode called a high frequency butterfly resonance mode.

A fourth resonance mode exists and is unintentionally created as a lower frequency butterfly mode, but it is not utilized in the parametric amplifier and will not be discussed in the following. The two zero-biased varactors in one of its end-ring are connected in a head-to-tail fashion so that its capacitance is modulated in the same direction during the application of pumping field. This end-ring creates the high frequency resonance mode, which is sensitive to an externally applied pumping field. The two vertical legs create a low frequency mode that is sensitive to a nuclear spin precession. Compared to a double frequency resonator, the quadruple frequency resonator has a larger length-to-diameter ratio, making $\omega_3 > 2\omega_1$. As a result, the idler signal at ω_2 is considerably far away from the MR signal at ω_1 , making it hard for these two signals to share the same resonance mode. This is also the reason why the higher-frequency butterfly resonance mode is introduced into the quadruple frequency resonator to sustain the idler signal. For efficient parametric amplification, this

butterfly mode resonance frequency should be approximated to $\omega_2 = \omega_3 - \omega_1$ by independent adjustment of the chip-capacitor C_5 .

C. Concatenated Parametric RF-Resonator

To further enlarge the longitudinal FOV, the quadruple frequency resonator is concatenated with the double frequency resonator. This concatenated detector uses a substrate as a cylinder that was 3D-printed with polyurethane. The cylinder was then wrapped by a copper-clad polyimide film that was etched to create the desired circuit pattern. The outside diameter (OD) of the concatenated cylindrical detector is 2.5 mm. The copper pattern wrapped around this polyurethane cylinder has a strip width of 0.5 mm, whose end-rings in the case of the double frequency resonator, are bridged by identical 32 pF diodes to create the ring resonance mode at 610 MHz (about twice the Hydrogen nuclei (^1H) Larmor frequency at 7 Tesla, T). The length of the double frequency resonator was empirically adjusted to 6.86 mm in order to make the horizontal dipole mode centered around 302 MHz. Similarly, a quadruple frequency resonator was constructed and designed according to the schematic diagram in Fig. 1. In this detector, all end rings are split by 22-pF chip capacitors or diodes to have ring resonance at ω_{30} 695 MHz, which is way above twice the Larmor frequency that is mandated for the double frequency resonator. To determine the proper length of the quadruple-frequency resonator for a dipole mode resonance frequency near 300.34 MHz (Larmor frequency of 7 T scanner), three lengths were tested to extrapolate its final desired end-ring distance. Initially, the first two detectors had end-ring distances of 10.64-mm and 10.8-mm, respectively. Each detector's scattering parameter S_{21} and the resonance curves for the horizontal dipole mode were measured. The low resonance frequency of the first and the second detectors were measured to be 306.8 MHz and 305 MHz, respectively. Subsequently, their inductance values were evaluated and plotted as a function of the length as shown in Fig. 2, with their linear relations given by:

$$y = 1.25 \times 10^{-6}x - 1.1 \times 10^{-9}, \quad (2)$$

where x is the length of the detector and y is the inductance.

Using (2), we then predicted that 11.2-mm is the appropriate end-ring distance for the final study. We built such a quadruple frequency resonator and it indeed had the desired resonant frequency around ω_{10} 299.1 MHz, which is close to the Larmor frequency of a 7 T MR-scanner. To sustain current flow independently at the idler frequency, a high frequency butterfly mode is created by splitting the center continuous ring with a 68-pF chip capacitor to have a resonance frequency at 399.8 MHz, which is close to $\omega_{30} - \omega_{10}$.

Fig. 3 shows a schematic diagram and a photo of the coated concatenated cylindrical resonator ready for MR experiments. These resonators are concatenated to each other having a gap distance of 1.35 mm, with the quadruple frequency resonator's end-ring with chip capacitors held adjacent to the double frequency resonator. The quadruple frequency resonator's scattering parameter (S_{21}) is measured by placing it above a dual pick-up loop probe, which is connected to a vector network analyzer. According to the solid curve in Fig. 4, the quadruple detector has a lower frequency at 299.1 MHz ($Q = 89$) in its passive state i.e., without pumping power. With the application of pumping power that is empirically

adjusted to ~ 0.5 decibel (dB) below the resonator's oscillation threshold, the dotted curve that shows up stably in Fig. 4 has a gain of ~ 18.5 -dB at the Larmor frequency of 300.34 MHz. The effective quality factor (Q_{eff}) of the resonator is found by dividing the center frequency (f_0) of the resonance peak with its bandwidth at -3 dB [31]. Evidently, with increased gain and effective quality factor, the bandwidth of the resonator decreases. The Q -factor of the active resonator is found to be $Q = 751$. Although the bandwidth at -3 dB decreases to 400 kHz, this reduced bandwidth is still much greater than the imaging bandwidth used in most MR experiments.

During an MR experiment, the correct operating condition for the resonator is established by empirically adjusting the frequency and amplitude of the pumping signal until the oscillation signal is observable as a single peak in the center of the MR spectrum [16]. To oscillate the quadruple resonator at Larmor frequency of 7 T MR scanner i.e. 300.34 MHz, the optimal pumping frequency is empirically adjusted to $\omega_3 = 701$ MHz, when the idler frequency is concomitantly determined by the following relation $\omega_2 = \omega_3 - \omega_1 = 400.66$ MHz. On the other hand, the optimal pumping frequency adjustment is not required for the double frequency resonator that can have sufficient gain at the Larmor frequency by simply setting the applied pumping frequency slightly above twice the Larmor frequency. Subsequently, the pumping power for each resonator is adjusted to 0.5 dB below the oscillation threshold, leading to 18.5-dB gain.

III. Results

A. Phantom Imaging Experiments

To test the performance of the concatenated parametric RF-resonator, the detector is first inserted in the center of a water phantom. A surface coil is placed externally underneath this detection object to receive amplified MR signals from the WAND while we use a volume coil (not shown) to excite a large volume uniformly for MR signal transmission. Both the WAND and the external receive coil are involved in MR image acquisition. The distance between the external surface coil and the WAND is about 22 mm. A large single-turn pumping loop is then inserted to feed the pumping power through an external RF-signal generator and positioned orthogonally to the main B_0 field (Fig. 5a). Subsequently, the whole assembly is inserted into a 7 T horizontal-bore MR scanner for imaging. During RF excitation, the WAND is decoupled from the excitation field that induced strong modulation of its constituent varactors. So, the varactor diodes in the WAND should detune the circuit during this phase. During MR signal acquisition, the resonator is activated as the varactors are modulated by the pumping signal for parametric amplification to improve the signal transmission efficiency to the external receiver (Fig. 5b).

Multi-slice fast-spin echo images are acquired with the following acquisition parameters: repetition time (TR) = 1s, echo time (TE) = 13.9 ms; FOV = 30×30 mm²; slice thickness = 0.8 mm; matrix size = 256×256 ; and imaging bandwidth = 50 kHz.

In Fig. 6, row 1 shows a series of transverse and longitudinal images acquired by the double frequency WAND (column 1, 2) and by the quadruple frequency WAND (column 3, 4). Simultaneous pumping is also applied to excite the two concatenated detectors to acquire a

longitudinal image (column 5). Row 2 shows relative sensitivity maps obtained by dividing images shown in Row 1 with images acquired by an external surface coil using the same acquisition parameters. As shown by the transverse profile in column 3, the quadruple frequency resonator has at least 3-fold sensitivity gain for regions up to 7-mm from its cylindrical surface and a sensitivity gain of at least 10-fold immediately adjacent to its surface. This distance is comparable to the corresponding effective range of the double frequency resonator in the transverse direction. According to the comparison of longitudinal profiles, the effective range of the quadruple frequency detector in the longitudinal direction (column 4) is about 26% larger than the double frequency resonator (column 2). When both the detectors are activated together, the effective range of both resonators overlay to create a larger longitudinal field-of-view (column 6).

B. In-vivo Imaging Experiments

Performance of our concatenated resonators is further tested for preclinical relevance by carrying out an *in-vivo* experiment on a 250-gm rat, which is approved and strictly carried out in accordance with the Institutional Animal Care and Use Committee (IACUC) at Michigan State University. The rat is anesthetized with 2% isoflurane inhalation and placed in a prone position inside the horizontal bore MR scanner. Placing this way will auto-enable the large portions of the rat's digestive tract, including esophagus and rectum, parallel to the main field B_0 . In addition, as we insert the concatenated resonator into the rat's esophageal region, its horizontal dipole resonance mode is perpendicular to its long axis, and the WAND will naturally get aligned at the correct orientation for MR signal detection. The detector is conveniently inserted into the esophagus connected to a 1.6 mm diameter 'Delrin' rod to adjust the detector's insertion depth in the esophageal region. Longitudinal slices around the aortic arch and common carotid artery regions are acquired with a flow-compensated gradient refocusing echo sequence (Fig. 7) using $TR = 5.3$ ms, $TE = 2.0$ ms, flip-angle (FA) = 10° , $FOV = 25.6 \times 25.6$ mm², slice thickness = 1 mm, matrix size = 192×192 , bandwidth = 50 kHz, and Number of Acquisition (NA) = 1. Compared to external detection (Fig 7a), an active amplifier can better observe the junctions (Fig. 7b) between the aortic arch and the supra-aortic great arteries, enabling better identification of vascular lesions at an early stage. Furthermore, one-dimensional intensity profiles plotted for both the images along the same horizontal line (yellow dashed line) show that the image acquired from the concatenated endogenous WAND has at least a five-fold sensitivity gain compared to the intensity profile obtained from the external detector alone. Such a detection ability will enable recognition of the images that could differentiate and identify vascular lesions.

IV. Discussion

In this work, concatenated RF-resonators with integrated parametric amplifier have been successfully constructed to demonstrate MR sensitivity enhancement capability over an extended region. All the resonators have a horizontal dipole resonance mode that can detect and re-emit amplified MR signals. By making the end-ring of each detector sensitive to a specific pumping frequency, it is possible to individually activate each detector and to optimize its respective gain level by the external pumping signal. When the end-ring's resonance frequency is approximately twice the Larmor frequency, the MR signal and the

idler signal can share the same horizontal resonant mode for efficient parametric amplification. When the ring resonance frequency is not near twice the Larmor frequency, however, it is necessary to create the additional butterfly mode to sustain current flow at the idler frequency and to enable efficient parametric amplification. Because of the high design symmetry of the quadruple-frequency resonator, the butterfly resonance can be independently adjusted by the chip capacitor on the center ring, without affecting the endring resonance or the horizontal resonance mode.

Despite of its extra design complexity, this quadruple frequency resonator is an improvement over previous versions of double frequency resonators [18], [32] because the ring resonance frequency of the double frequency resonator had to be set to around twice the MR frequency. The fixed ratio between the ring-resonance and the horizontal resonance mandated a relatively small dimension ratio between the resonator's length over diameter, limiting its longitudinal field-of-view [18], [20]. In comparison, the quadruple frequency resonator can receive a pumping signal at a higher frequency, enabling a more flexible choice of the detector's length. In principle, it is possible to cover an extended FOV with an elongated detector. However, if the length of a single detector is too large, its sensitivity gain could also be compromised when the excitation B_1 -field produced by unit current is reduced. Alternatively, it is also possible to cover an extended FOV with multiple concatenated resonators that are sensitive to the same pumping frequency. But because each resonator will inevitably have slightly different coupling efficiency with the pumping field, a single pumping signal will not allow wireless adjustment of individual detectors for optimal performance. Therefore, to get an extended FOV without sacrificing the performance of individual detectors, we concatenate the double and quadruple frequency resonators together, so they can be individually manipulated by pumping signals separately provided by different frequency synthesizers.

Besides the additional requirement for an extra frequency synthesizer, one limitation of the current study is the small signal void between adjacent resonators. This limitation could be overcome by minimizing the separation gap between adjacent resonators or even partially overlapping them while minimizing their mutual coupling. Active work is performed along this direction.

V. Conclusion

Concatenated design of cylindrical parametric RF-resonators has been implemented to demonstrate resonators' individual manipulability by the external pumping signal for sensitivity enhancement. Both the double and quadruple frequency resonators are concatenated together to effectively increase the longitudinal field-of-view. In contrast to a single-element double frequency resonator, where the idler and the MR signal frequencies share the common mode, the quadruple frequency resonators have idler current flowing in the butterfly mode that can be independently adjusted. The double frequency resonator is mandated to have a small length-to-diameter ratio, because of the requirement that the pumping frequency need to be twice the detection frequency. In contrast, the quadruple frequency resonators can surpass the smaller length-to-diameter ratio as it can be pumped at a frequency other than twice the detection frequency. The ability to individually manipulate

each resonator by the pumping frequency also make it feasible to effectively concatenate them to increase the overall field-of-view along the longitudinal direction. Thus, this design has enlarged the longitudinal field-of-view without compromising the effective axial detection range when compared to previous versions of single-element cylindrical double frequency resonators. We anticipate that our concatenated parametric resonators would be more beneficial when it is necessary to characterize deep-lying lesions over an extended region, e.g. along the aorta.

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Biographies

Roshan Timilsina received the B.S. and M.S. degrees in physics from Tribhuvan University, Kathmandu, Nepal, in 2009 and 2012, respectively. In 2014, he joined the Department of Physics, University of Akron, OH, USA to pursue his international MS degree in physics.

He is currently working toward his Ph.D. degree in biomedical sciences: medical physics at Oakland University. As part of the degree requirement, he is conducting his research in the Department of Radiology, Michigan State University in conjunction with Department of Physics, Oakland University designing and fabricating wireless amplified radio frequency (RF) coils to develop advanced detection sensitivity and accuracy for next generation MRI diagnosis.

Chunqi Qian received his BS degree in chemistry from Nanjing University and his Ph.D. in physical chemistry from the University of California, Berkeley, in 2007. Following postdoctoral trainings at the National High Magnetic Field Laboratory and the National institutes of Health, he became a principle investigator in the department of Radiology in Michigan State University. He is focusing on the development of advanced imaging technology for biomedical research.

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Fig. 1.

Illustrations of various resonance modes in a quadruple frequency resonator. (a) In the ring resonance mode, the direction of the arrows show the current flow inside the two end rings that creates the high frequency resonance modes sensitive to a longitudinal pumping field. (b) In the horizontal dipole resonance mode, the direction of the arrows show current flow through the vertical legs, from which this lower-frequency resonance mode is created to be sensitive to nuclear magnetization precessing in the horizontal plane. (c) In the butterfly mode one, the counter-rotating currents in the upper and lower halves of the circuit primarily pass through the continuous ring in the center plane. (d) In the butterfly mode two, the counter-rotating currents in the upper and lower halves of the circuit primarily pass through the chip capacitor C_5 in the center plane.

Fig. 2.

Inductance as a function of length for three different lengths of quadruple frequency resonators. The graph shows the linear relationship between their inductance and length, where the slope is 1.250×10^{-6} H/m and y-intercept is 1.100×10^{-9} H. The three resonators considered had lengths of 10.64 mm, 10.8 mm, and 11.2 mm.

▶ : Varactor

Fig. 3.

Concatenated RF Resonator. (a) A schematic diagram of a cylindrically symmetric concatenated WAND. The detector is made of a copperclad polyimide etched film mounted on a polyurethane cylinder with 2.5-mm outside diameter (OD), and 1.72-mm inside diameter (ID). The ending distance is 11.2 mm for quadruple-frequency resonator and 6.86 mm for double frequency resonator. There is also a 1.35-mm gap between these two resonators. In the quadruple-frequency resonator, C_1 and C_2 are 22-pF chip capacitors. C_3 and C_4 are varactor diodes (MA2737600L, Panasonic) with zero-biased capacitance of 22

pF. The center chip capacitor C5 is 68 pF. In the double frequency resonator, all varactor diodes (BAS3005A, Infineon) on end-rings (C_1 , C_2 , C_3 , C_4) have zero-biased capacitance of 32 pF. (b) A photo of the concatenated cylindrically symmetric resonator coated by a thin layer of medical grade epoxy and a heat shrink tubing.

Fig. 4.

The S_{21} curve of a quadruple frequency resonator. The solid curve is when the resonator is in its passive state without application of the pumping field, whereas the dashed curve is in its active state with the application of pumping power at ~ 0.5 dB below its oscillation threshold.

Fig. 5.

Schematics for the experimental setup for animal imaging using the WAND. (a) The rat is placed in a prone position and the WAND is inserted through the esophagus near the aorta arch. External pumping signal is supplied via the RF signal generator through a single turn inductor loop to the WAND. (b) Zoom-in view of the WAND in Fig. (a). The amplified MR signals from the WAND is received by the external receiving coil placed underneath the rat and to the MR receiver for further processing.

Fig. 6.

2D fast spin echo MR images obtained by the concatenated RF-resonators in a water phantom. Column 1, and Column 2 show the transverse and longitudinal images (row 1) and their relative sensitivity maps (row 2) when only the doubly frequency resonator is activated. Column 3, and column 4 show the transverse and longitudinal images and their relative sensitivity maps when only the quadruple frequency resonator is activated. Column 5 shows the MR image and the relative sensitivity map when both the detectors are activated.

Fig. 7.

A longitudinal gradient-echo slice acquired around the region of the aortic arch and common carotid artery of a rat heart in the absence of pumping power using an external detector only (a), and in the presence of pumping power with the use of a concatenated endo-esophageal detector (b). The parameters used for image acquisitions are TR/TE=5.3/2 ms, 10° flip angle (FA), 25.6 × 25.6 mm² FOV, 1-mm slice thickness, 192 × 192 matrix, and NA = 1. Row 2 are the 1D-normalized intensity profiles plotted for each spot corresponding to each image

in row 1 along the horizontal yellow dashed line and highlighted within vertical red dashed lines.

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