

Metamaterial MRI-based Surgical Wound Monitor

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Abstract – An implantable sensor for monitoring wound healing after bowel reconstruction is demonstrated. The sensor consists of a pair of magneto-inductive ring resonators, designed for mounting on a biofragmentable anastomosis ring and inductively coupled to an external coil to give a local increase in signal-to-noise ratio near an annular wound during ¹H magnetic resonance imaging. SNR enhancement is confirmed using thin-film prototypes operating at 3T.

I. INTRODUCTION

Anastomosis is a surgical procedure involving suturing, stapling or compression rings, used to restore continuity after bowel tumour resection. Despite continuing advances, leakage remains a potentially devastating complication, with a high (6%-22%) mortality [1]. The cause is poor blood supply (ischemia), aggravated by mechanical tension and tissue damage from radiation therapy. Leaks are generally identified using computed tomography (CT), but only when fully developed and the prognosis is poor. Consequently there is a need for sensing modalities capable of detecting conditions likely to lead to a leak. One possibility is magnetic resonance imaging (MRI) or spectroscopy (MRS). ¹H and ³¹P MRS have been used to monitor colonic ischemia, based on the increase in lactate or the reduction in adenosine triphosphate peaks relative to phosphocreatine [2, 3]. Unfortunately, due to body noise, MRS has a low signal-to-noise ratio (SNR) using external coils, which leads to poor results in the abdomen [4]. However, SNR may be locally increased using internal coils with a small field-of-view (FOV) for body noise, coupled inductively to external interrogation coils [5]. Here we demonstrate a metamaterial sensor based on magneto-inductive ring resonators [6, 7]. The sensor has a FOV matched to annular wounds, and is designed for eventual implantation in a biofragmentable anastomosis ring (BAR) [8].

II. DESIGN

Figs. 1a-c show a BAR, which consists of two connecting rings of biodegradable polymer designed to provide compression between well-perfused sections of bowel lumen for around two weeks, before natural excretion. To match the surgical wound, the desired FOV of the implanted coil is annular (Fig. 1d). A magneto-inductive ring (Fig. 1e) can provide a suitable FOV, and due to its lack of physical connections is also fragmentable. However, to ensure decoupling from the B₁ field during excitation, we propose a coupled pair of rings operating on their lowest-order antisymmetric spatial mode, with each ring mounted in a separate half of the BAR (Fig. 1f).

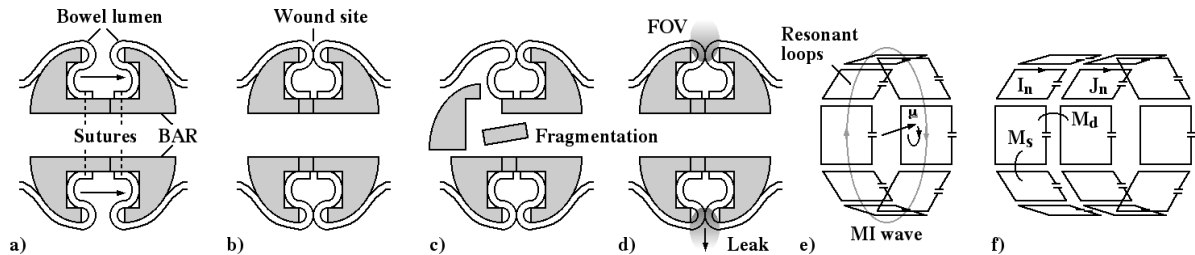


Fig. 1 a) - c) Operation of a BAR; d) desired field-of-view; e) and f) single and double magneto-inductive rings.

Each ring is made from elements with self-inductance L and capacitance C , with nearest-neighbour intra-ring mutual inductance M_s and inter-ring mutual inductance M_d . For two infinitely long, lossless coupled MI waveguides with loop currents I_n and J_n the circuit equations at angular frequency ω are:

$$ZI_n + Z_{Ms}(I_{n-1} + I_{n+1}) + Z_{Md}J_n = 0 ; ZJ_n + Z_{Ms}(J_{n-1} + J_{n+1}) + Z_{Md}I_n = 0 \quad (1)$$

Here $Z = j\omega L + 1/j\omega C$, $Z_{Ms} = j\omega M_s$, and $Z_{Md} = j\omega M_d$. Assumption of wave solutions $I_n = I_0 \exp(-jnka)$ and $J_n = J_0 \exp(-jnka)$ yields the dispersion equation for the symmetric ($J_n = I_n$) and anti-symmetric ($J_n = -I_n$) branches as:

$$\omega/\omega_0 = 1/\sqrt{\{1 + \kappa_s \cos(ka) \pm \kappa_d/2\}} \quad (2)$$

Where k is the propagation constant, a is the element spacing and $\kappa_s = 2M_s/L$ and $\kappa_d = 2M_d/L$ are intra- and inter-ring coupling coefficients. For ring resonance, the propagation constant must satisfy $ka = 2\pi\mu/N$ where N is the number of elements in each ring and $\mu = 0, 1 \dots$ is the mode number [6]. For $N = 8$ (for example) there are ten modes overall, five symmetric and five anti-symmetric. Fig. 2a shows the dispersion diagram for typical values of $\kappa_s = -0.2$ and $\kappa_d = -0.1$. In the planar case ($\kappa_s < 0$), backward waves propagate in both rings. However, the lowest-order ($\mu = 1$) anti-symmetric resonance, arrowed, is close in frequency to other resonances, suggesting poor rejection of uniform B_1 fields. If, however, the intra-ring coupling κ_s is made positive, forward waves are obtained, with improved separation of the lowest order anti-symmetric resonance as shown in Fig. 2b. To achieve this, adjacent elements are merely required to overlap in the quasi-axial configuration shown in Fig. 2c.

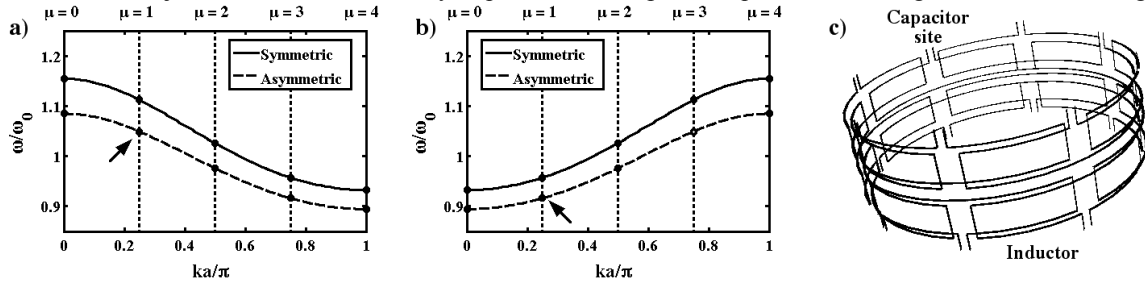


Fig. 2 Dispersion diagrams for negative κ_d and a) negative and b) positive κ_s ; c) coupled rings with positive κ_s .

III. FABRICATION AND EXPERIMENTAL EVALUATION

Coupled magneto-inductive ring resonators with the layout of Fig. 2c were constructed with a diameter of 34 mm and an axial length of 25 mm. Each ring was composed of eight rectangular copper inductors measuring 4 x 24 mm on flexible plastic substrates. The circuit was mounted on an annular plastic cylinder and made resonant with non-magnetic capacitors, with the lowest order anti-symmetric mode tuned to 127.7 MHz, the Larmor frequency for ^1H MRI in a 3T magnetic field (Fig. 3a). The sensor was then encapsulated in epoxy resin and embedded in a 70 mm diameter tissue-simulating agar gel phantom doped with NiCl_2 and NaCl (Fig. 3b). MRI was carried out in a GE MR 750 Discovery clinical scanner, with the test phantom between two cuboid loading phantoms and using the system body coil for excitation and an 8-element array coil for reception (Fig. 3c).

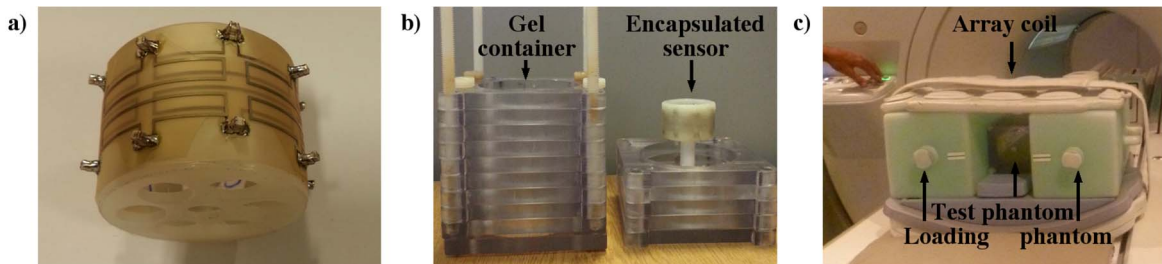


Fig. 3 a) Sensor before encapsulation, b) gel phantom, and c) arrangement for imaging.

Imaging was carried out using a spin-echo sequence with flip angle 90° , excitation time 15 ms, repetition time 700 ms and slice thickness 2 mm. Figs. 4b-d show axial slices across the sensor at the locations shown in Fig. 4a. In each case, the outer black regions represent the gel boundary, while the light grey regions indicate the normal response of the array coil. The central black annulus shows the location of the coil mount, while the six grey circles in Fig. 4d represent holes in its base to allow gel flow. Fig. 4c shows a null in sensitivity between the two rings, aligned with the spatial zero of the anti-symmetric mode. However, Figs. 4b and 4d show the desired circumferential increase in brightness due to local amplification of the MRI signal by the resonator.

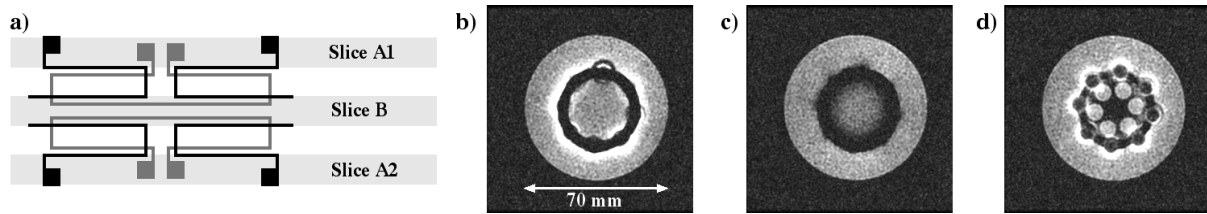


Fig. 4. a) Slice positions and b) axial slice images of coil test phantom.

Fig. 5a shows a theoretical sensitivity map calculated using reciprocity for a single eight-element MI ring formed from semi-infinite wires, which has a similar local enhancement. Fig. 5b shows a quantitative comparison of sensitivity obtained by matching a signal baseline due to the array coil and the theoretical enhancement due to the sensor to normalized experimental data from slice A1 along the line $y = 0$. There is reasonable agreement, and the ratio of the maximum to the baseline suggests a local enhancement factor of 4.

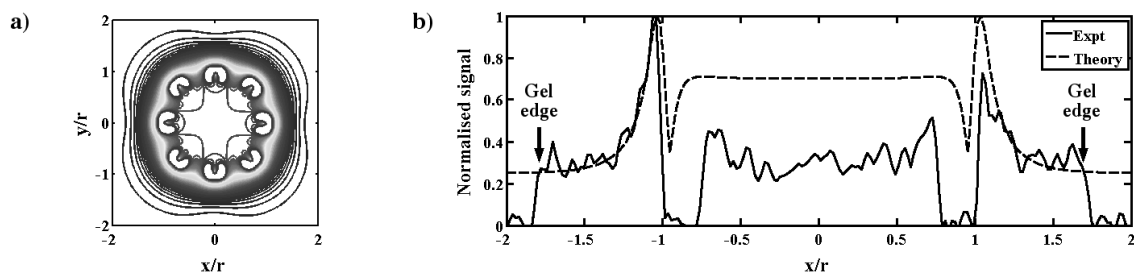


Fig. 5 a) Theoretical sensitivity map of 8-element ring; b) linear variation of sensitivity across the centre of slice A1.

IV. CONCLUSIONS

An internal sensor for surgical wound monitoring by MRI has been demonstrated, based on a pair of magneto-inductive rings. The intra-ring coupling coefficients are chosen to maximize mode separation, and the sensor operates on the lowest order anti-symmetric spatial mode to minimize coupling to the B_1 field during excitation. ^1H phantom imaging experiments at 3T show a peak circumferential signal enhancement of four.

V. REFERENCES

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