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Thin-Film Detector System for Internal Magnetic Resonance Imaging

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ABSTRACT

A detector for *in vivo* internal magnetic resonance imaging (MRI) is demonstrated based on a thin-film RF resonator and a thin-film cable. Each component is constructed on a flexible sheet and mounted on the outside of a catheter, leaving its internal lumens free for clinical use. Space constraints require the sheet to be extremely thin ($< 100\ \mu m$). Cable formats are compared, and thin-film cables with $\approx 50\ \Omega$ impedance at low frequency are formed as a microstrip with a periodically patterned ground, using copper conductors on polyimide substrates. Resonant detectors are also formed on polyimide from multi-turn electroplated copper coils and integrated parallel plate capacitors, which use the substrate as an interlayer dielectric. Methods are developed for obtaining capacitor values for matching and tuning, and compensating for loading. The detector and cable are linked to form a two-metre-long printed detection system and ^1H MRI is demonstrated at 1.5 T using *in vitro* liver tissue.

1. Introduction

Small resonant RF detectors have many applications for *in vivo* internal magnetic resonance imaging (MRI), for example for rectal, biliary and arterial imaging. Although small coils generally have low Q factors, this disadvantage is mitigated by the increase in signal-to-noise ratio obtained from close coupling to the signal source [1]. Suitable coil arrangements include single-or multi-turn loops [2–4], parallel conductor transmission lines [5] and opposed solenoids [6,7], constructed by hand winding or from rigid printed circuit boards (PCBs). Compact alternatives for vascular imaging include the loopless catheter antenna [8]. A similar range of coils has been used for catheter tracking [9,10].

In each case, the need for matching and tuning has limited widespread application. Several well-known circuits (for example, shunt or series capacitive matching) may be used. Unfortunately, despite advances in modelling methods that allow 3D coils, skin effects and material losses, it is difficult to estimate coil inductance and resistance accurately. Consequently, capacitor values must be determined experimentally. Multiple capacitors may be needed, and circuits may be re-soldered many times. The end product is acceptable for large coils, but cannot achieve the low cost, small form factor and reproducibility needed for mass deployment. One alternative is to locate the components remotely, using a $\lambda/2$ length of cable [11]. However, the strategy is difficult to implement with long catheters.

Microfabrication can improve the situation, since it yields repeatable inductance and resistance. Electroplated spiral coils have been formed on GaAs [12,13], Si [14,15] and glass [16,17]. Microfabricated Helmholtz coils have been constructed [18,19], solenoids have been fabricated on capillaries [20–22], planar coils have been integrated with microfluidics [23,24], and pre-amplifiers have been incorporated [25–27]. More recently, attention has turned to flexible plastics such as polyimide and polyether-ether-ketone [28] and polytetrafluoroethylene [29], which are more suitable for *in vivo* use. Some attempts have been made to integrate capacitors, using coplanar electrodes [30] or double-layer windings [31]. However, no convincing solution to matching has been demonstrated.

We ourselves have demonstrated high-resolution MRI using catheter-mounted flexible microfabricated coils with discrete capacitors [32]. The target application was an endoscopically delivered detector for *in vivo* imaging of the bile duct, with the aim of early detection of cholangiocarcinoma. Because operable tumours must typically be less than1 mm in size, imaging must be carried out with sub-millimetre resolution. The length and diameter of the common bile duct range from 5-10 cm and from 3-8 mm, respectively; however, the duct is normally heavily constricted near any lesion. A 60 mm long coil capable of mounting on an 8 Fr (2.7 mm dia) catheter was developed, which did indeed have suitable resolution.

However, in addition to the matching difficulties described above, little attention was paid to clinical use. For example, the

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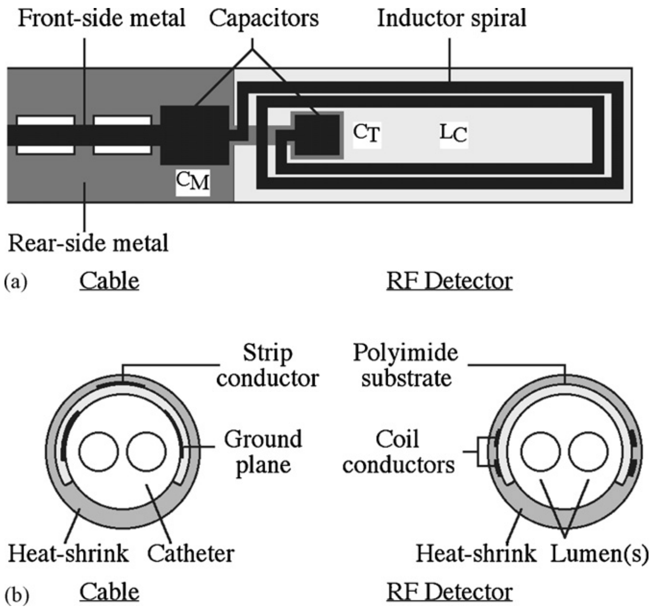

Figure 1. Thin film RF detection system: a) plan view and b) integration on catheter.

capacitors were vulnerable to mechanical damage in a side-opening gastroscope, and the need for joints made it difficult to seal the assembly for use in a wet environment. The sub-miniature coaxial cable used to transmit the detected signal back along the catheter blocked one of the internal lumens, which are typically required for use with a guide-wire or an actuation wire or for injection of contrast agent.

In this paper, we attempt to provide a solution in the form of a thin-film detector with integrated shunt matching capacitors and a thin-film output cable as shown in Figure 1a. The whole structure may be fabricated lithographically from non-magnetic materials and wrapped around a catheter as shown in Figure 1b, avoiding the need for additional components, simplifying sealing and leaving the catheter free for its original use. Except near the coil ends, the configuration is similar to a two-turn planar coil with stacked windings, with a corresponding field of view. The target application here is biliary imaging, which may require data to be acquired for lengths 50–60 mm along the duct. Similar approaches could be used for vascular imaging.

Operation of the system is shown in the equivalent circuit of Figure 2a. Here, a source V_S representing the signal induced by nuclear magnetic dipoles in a resonator based on a coil with inductance L_C and resistance R_C is to be matched to a load R_L at the angular frequency ω_S of the MRI system using matching and tuning capacitors C_M and C_T. Generally, R_L should be matched to the series sum of R_C and the sample loading R_S. However, assuming that R_S is small (as shown experimentally later) and that the cable impedance Z_0 matches R_L, the system operates as shown in the simplified equivalent of Figure 2b.

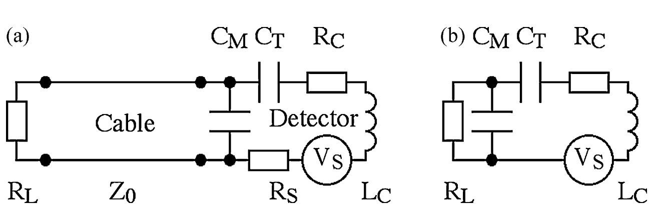

Figure 2. a) Electrical equivalent circuit and b) simplified equivalent.

Key to successful operation is development of impedance-matched cables and resonant detectors containing suitable matching circuits. Geometric constraints and the need for flexibility make it difficult to achieve the desired impedance using either a microstrip [33] or a coplanar waveguide [34]. Here we use a microstrip with a periodically patterned ground plane. This approach is commonly used in photonic bandgap devices [35], but it can also modify low-frequency impedance (a factor that was previously ignored in favour of filter applications). We use a simple resonant detector with integrated capacitors that can be fabricated using compatible processing, and demonstrate a convergent method of identifying the component values needed for matching and tuning. Combining these elements we demonstrate external ^{1}H magnetic resonance imaging at 1.5 T using a 2 metre-long catheter based detector, constructed entirely from thin film components.

A design for an impedance-matched thin-film cable is presented in Section 2, and fabrication and measurements of transmission characteristics in Section 3. A design for a compatible thin-film coil with integrated capacitors is presented in Section 4, and fabrication and matching and tuning are described in Section 5. Preliminary MRI experiments are described in Section 6 and conclusions are drawn in Section 7.

2. Thin-film cable design

In this Section, we consider possible cable formats, including microstrip, coplanar waveguide and periodically patterned microstrip, for an application such as a catheter-based MRI detector requiring very thin flexible layers of metals and dielectric.

2.1. Microstrip

Figure 3a shows one obvious possibility, a microstrip consisting of two conducting layers separated by an insulator of relative dielectric constant ε_r. One conducting layer is patterned into a strip of width w, and the thickness of conductor and insulator are t and h respectively. The characteristic impedance Z_0 can be estimated using Wheeler's formulae [33]:

$$Z_0 = Z_{FS}\, c_1 \log_e\{1 + c_2[c_2 c_3 + \surd(c_2^2 c_3^2 + c_4)]\} \quad (1)$$

Here $Z_{FS} = \surd(\mu_0/\varepsilon_0)$ is the impedance of free space, and:

$$\begin{aligned} &c_1 = 1/\{2\pi\surd[2(1+\varepsilon_r)]\} \quad c_2 = 4h/w_{eff} \quad c_3 = (14 + 8/\varepsilon_r)/11 \quad c4 = \pi^2(1 + 1/\varepsilon_r)/2 \\ &w_{eff} = w + t\, c_5 \log_e\{4e/\surd(c_6^2 + c_7^2)\} \\ &c_5 = (1 + 1/\varepsilon_r)/2\pi \quad c_6 = t/h \quad c_7 = 1/\{\pi(w/t + 11/10)\}^2 \end{aligned} \quad (2)$$

These expressions give the typical results shown in Figure 3a. Here, we have assumed t = 35 μm and ε_r = 3.5, to model a Cu-polyimide-Cu trilayer. Two curves are shown, for h = 25 μm and h = 50 μm, and 50 Ω impedance is only obtained for very small (< 100 μm) strip widths, with w reducing as h decreases. The explanation is the small separation of the conductors, which results in low inductance and high capacitance per unit length. Reliable lithographic fabrication in long lengths is therefore likely to be difficult. Larger inductance and/or smaller capacitance are required, but using moderate (w ≈ 1 mm) strip widths.

2.2. Coplanar waveguide

Figure 3b shows a second possibility, a coplanar waveguide (CPW) consisting of a ground-signal-ground arrangement on a dielectric backing. Z_0 is found by evaluating the capacitance C_p per