Defence Science Journal, Vol 43, No 3, July 1993, pp 259-262

Development of RF Coil for ^{31}P *in-vivo* NMR Spectroscopy

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ABSTRACT

RF receiver coils are very important parts of an NMR System. The design of these coils is very critical and has a dramatic effect on the SNR of the NMR signal and are generally developed in TRA/REC mode. This paper reports the development of a 3.5 cm TRA/REC 26 MHz RF coil for ^{31}P spectroscopy of small organs like thyroid. The coil is small in size, fits well in the neck for thyroid spectroscopy and is successfully working with the 1.5 tesla whole body Superconducting NMR System available at INMAS.

INTRODUCTION

The RF coil generates oscillatory fields that are used to excite the nuclei and detect the signals in NMR experiment. It is a good practice now to use an optimised coil in shape and size that just accommodates the part of the body under examination. Such RF coils which are placed over the region of interest and specially designed in shape and size to improve the performance are termed as surface coils[1]. In order to optimise the performance of the coil, it is essential to have the best filling factor which expresses the ratio of the total coil volume to that occupied by the patient. The two distinct advantages of surface coils over other conventional coils are improvements in the spatial resolution and the two to ten fold improvement in the signal-to-noise ratio (SNR). Essentially surface coils are equivalent to one turn solenoids; they have a field of view that is approximately a hemisphere with the same radius of the coil and their sensitivity falls off with distance from the centre. These coils produce inhomogeneous B1 field in contrast to a conventional coils which produce homogeneous field for uniform excitation. In spectroscopy, the B1 inhomogeneity of the coil is used as a method of spatial localisation[2-4] of the NMR signal. One of the major source of noise in clinical NMR is the patient[5,6]. The controlled region of sensitivity possible with surface coil results in further improvements; the thermal noise is eliminated from parts of the body outside the sensitive region, and the artifacts due to the patient motion, respiration, etc are also reduced.

2. CIRCUIT DETAILS AND DESIGN CONSIDERATIONS

The SNR in magnetic resonance receiver coil is represented by the equation

$$SNR = KnM_op \ (QW_o V_c/4FkT_cf)^{1/2} \qquad 1)$$

where Q is the quality factor of the coil loaded, n is the filling factor of the coil, M_o is the nuclear magnetisation, p is the proximity factor, K is a constant which depends on the coil geometry, W_o is the resonance frequency, V_c is the coil volume, F is the noise figure of the pre-amplifier, k is the Boltzmann constant, T_c is the temperature of the probe, and f is the receiver bandwidth. The quality factor of the coil is defined as

$$Q = \frac{f}{[a + b\,(f)^{1/2} + df^2]} \qquad (2)$$

where a is proportional to the dc current resistivity of the coil material, b is the RF resistivity of the material

Received 10 August 1993

and d is dependent on the coil loading factor. At higher frequencies (currently used in NMR imaging and spectroscopy) the coil loading factor dominates (the denominator) and consequently determines the SNR of these coils. The filling factor, coil loading factor and hence the Q factor of the coil have to be critically considered while taking up coil design for specific applications.

The circuit designed consists of essentially a parallel tuned circuit[7] with additional capacitors for coupling. The efficient power transfer takes place only if the coil circuitry matches with that of the cable and the RF amplifier. Cables usually have a characteristic impedance of 50 Ω and it is therefore important to match this coil circuit which appears to behave as a 50 Ω resistance. The characteristics of the coil itself are different from this and therefore are cannot simply connect the coil across the ends of the cable. This is achieved by the circuit in Fig. 1.

The tuning capacitor C_t of the coil, which has an inductance L and resistor r constitutes a tuned circuit which has a resonance frequency

$$W_o = 1/(LC_t)^{1/2} \tag{3}$$

At this frequency, particularly large magnetic fields can be generated at a given input power. However the tuning capacitor in itself does not ensure efficient power transfer because the circuit will still not appear as a 50 Ω resistance; for this an additional matching capacitor C_m is required. Prior to start the calibration, the coil is to be tuned and matched, i.e., the values of C_m and C_t are such that the coil behaves like a 50 Ω resistance at the resonance frequency, i.e., 26 MHz. Coil tuning was done with the help of an RF bridge and is achieved by adjusting C_t and C_m until a null is observed from the bridge. Figure 2 shows the prototype of the indigenously developed coil.

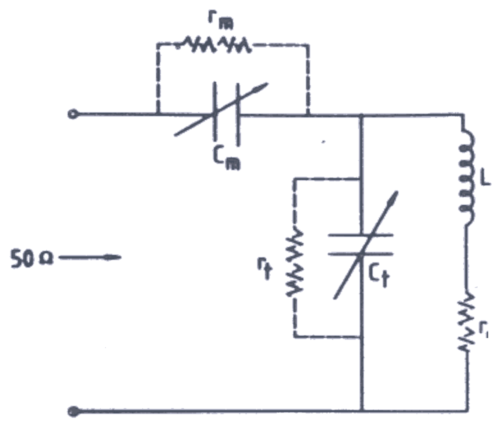


Figure 1 Parallel resonance circuit with an additional matching capacitor C_m.

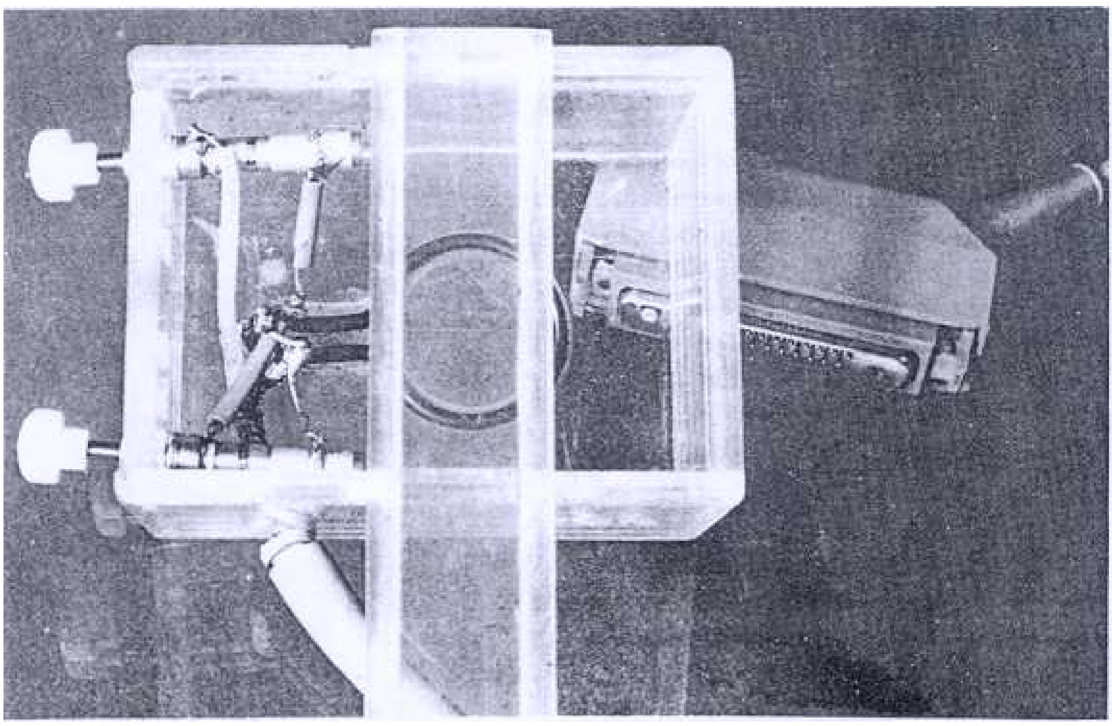


Figure 2. Prototype of the indigenously developed coil.

3. MATCHING OF COIL

Matching involves the optimum transfer of energy into or out of the circuit. The Q of the coil is lowered by the addition of an external load of 50 Ω and an extra reactance is necessary to effect this matched condition. Thus somehow, the external load of 50 Ω must have its resistance changed so that it introduces a value r in series with the resistance already present. Consider a 50 Ω resistance in series with matching capacitor C_m in (Fig. 1) whose reactance $1/WC_m \gg 50\ \Omega$. Then effecting a series to parallel transformation the combination is a resistance R'

$$R' = 1/50W_o^2 C_m^2 \tag{4}$$

in parallel with a small capacitance and the parallel tuned circuit. However, if we make $R' = QW_o L$, a parallel to series transformation using the inductor shows that we have introduced an extra series resistor r. The condition of matching then will be

$$R' = (1/50W_o^2 C_m^2) = QW_o L = (Q/W_o C_t) \tag{5}$$

Therefore,

$$(Q/W_o C_t) = (1/50W_o^2 C_m^2) \tag{6}$$

or

$$C_m = (C_t/50W_o Q)^{1/2} \tag{7}$$

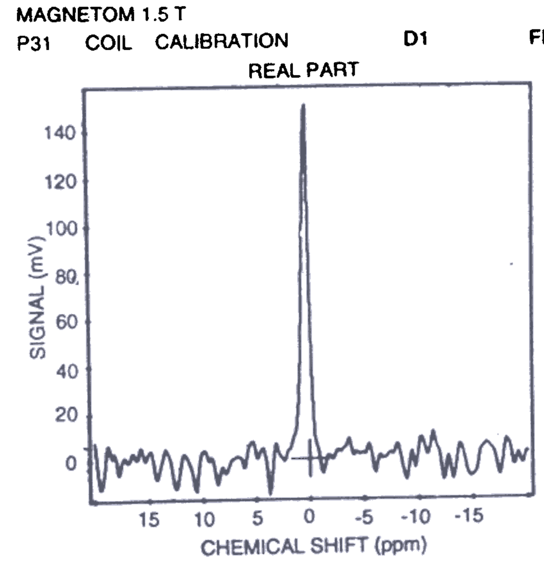
The capacitors used were high Q (>3000) with teflon dielectric to ensure that they do not introduce any additional noise in the tuned circuit. The capacitors should be able to withstand the high voltages used in FT NMR and most importantly that they should be made of nonmagnetic material. The tuning capacitor should be placed as close as possible to the RF coil in order to minimise the resistive losses in the leads of the coil. The earth connection to the circuit was made by means of copper foil that surrounds the coil.

4. CALIBRATION OF THE COIL FOR 90 DEGREE PULSE

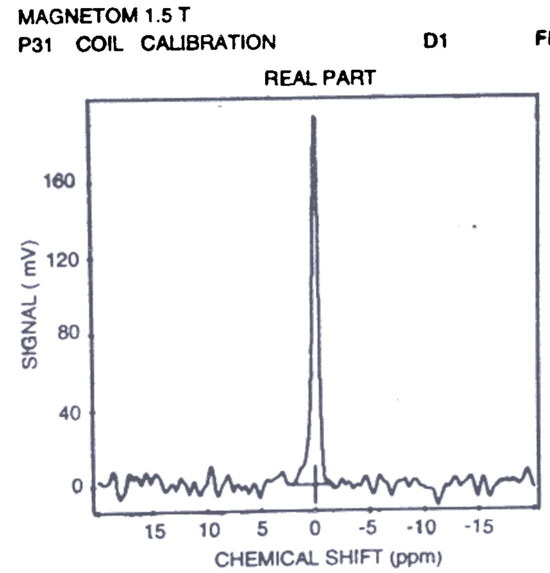
After tuning the coil at 26 MHz, i.e., ^{31}P frequency, the coil was calibrated for 90° transmitter pulse. To do this a disc phantom with 0.1 mM orthophosphoric acid was made and spectra were acquired using FID sequence. TRA voltage was gradually increased from 2 V onward in steps of 2 V till the amplitude of signal becomes maximum and then decreases again. For our coil, it was seen that at TRA voltage of 18 V the signal amplitude was maximum showing a 90° pulse at 18 V.

5. RESULTS AND DISCUSSIONS

For efficiency evaluation the phantom spectra were taken using this coil and compared with that of commercially available spertra. Figure 3 shows are the spectra taken by the coil designed and developed at



(a)



(b)

Figure 3. ^{31}P spectre of the phantom using, (a) commercially available coil, and (b) indigenously developed coil.

Institute of Nuclear Medicine and Allied Sciences (INMAS) and commercially available coil, respectively. It is seen that the SNR of the indigenous coil is better than the coil commercially available, possibly because of the use of high Q capacitors (Teflon dielectric). The Q factor of the indigenous coil was found to be greater than 300 at 26 MHz. It is useful to note that for a high quality tuned circuit C_m should be smaller than C_t by a factor of about 5 or more, and this factor decreases as the quality of the circuit declines. Figure 4 shows the spectra acquired using this coil from human calf muscle. All the phosphorous metabolites are clearly seen in the spectra with phosphocreatine peak as the reference.

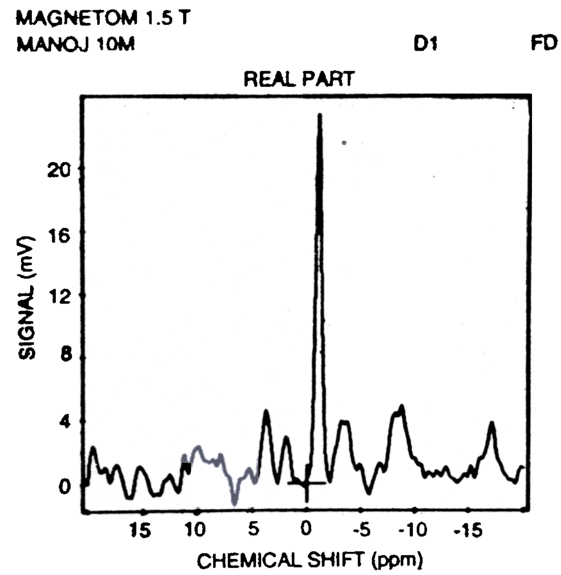


Figure 4. **^{31}P spectra of the calf muscle of a human volunteer. Peak assignments at (i) β-ATP, (ii) α-ATP, (iii) γ-ATP, (iv) Pcr, (v) phosphodiester, and (vi) inorganic phosphate.**

The coil developed at INMAS is working satisfactorily on the 1.5 tesla Superconducting NMR System at INMAS for *in vivo* ^{31}P spectroscopy. Attempts are being made to further improve the coil performance for NMR applications on patients.

ACKNOWLEDGEMENTS

Authors acknowledge with thanks the support given by Shri BS Rawat and Workshop staff of INMAS for fabricating the device.

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