Automated Tuning and Matching of a 16-Channel TEM Transmit-Only Array used in Conjunction with a 32-Channel Receive-Only Loop Array for MR Cardiac Imaging at 7 Tesla.

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Abstract — Radiofrequency (RF) coils are antenna-like devices used in magnetic resonance imaging (MRI) to inductively excite and receive the nuclear magnetic resonance (NMR) signal in anatomy. Recently, coil arrays have shown significant advantages over large single channel volume coils at ultra-high magnetic field strengths. However, to maximize the full potential, each individual coil within the array needs to be “tuned and matched” to the anatomy (load) prior to imaging. Tuning and matching can be a tedious and time consuming process, and has been cited as a problem that precludes ultra-high field strength MRI from becoming a clinical reality. Here, we have developed an electromechanical solution. A 16-channel transmit-only TEM array was designed and fitted with piezoelectric actuators mechanically linked to sliding cylindrical tuning and matching capacitors to automate the tuning and matching process, based on the coil’s reflection coefficient.

Index Terms — Impedance matching, Magnetic resonance imaging, Microstrip arrays, Piezoelectric actuation, 7 tesla.

I. INTRODUCTION

Magnetic Resonance Imaging (MRI) is becoming a common diagnostic tool in hospitals. The main magnet field strength (B0) of clinical MR usually ranges from 0.2 T to 3.0 T, while research magnets up to 9.4 T have been created for human subjects [1]. One of the major benefits of higher B0 field strengths is the increase of signal to noise (SNR); the SNR correlates in approximately a linear fashion with field strength.

Imaging the human body at ultrahigh fields, however, poses substantial challenges. One of the main difficulties comes from the spatial inhomogeneities and reduced efficiency resulting from the constructive and destructive interferences between complex B1 vectors. At 297 MHz (the precession frequency of the hydrogen nucleus (the Larmor frequency) at 7T) the RF wavelength in body tissues are comparable to or shorter than the object of interest resulting in non-uniform excitation and receive patterns due to complex RF field interferences. These interferences reduce both the RF transmit efficiency and homogeneity, potentially increasing localized power deposition as measured by the specific absorption rate (SAR). The ability to transmit through multiple independent channels (i.e. an array of coils) [2] provides the most general and flexible solution to these problems using techniques like static B1 shimming [3-5] or spatially tailored RF pulses.

For an MR coil or an array of coils to be most effective, each coil must be tuned to the Larmor frequency and matched to a 50Ω transmission line. This can be difficult as the resonance frequency and quality factor of a coil will change when it is loaded to different body weights, shapes, and tissue compositions. If a coil is not on resonance and matched in the loaded condition, two major effects will occur. First, the transmit coil will have reduced efficiency and second, the radiated power decreases because the reflected power increases.

In the research setting, the scientist or engineer is often required to manually tune and match the array via variable capacitors or inductors. This, however, can be a lengthy and time consuming process and has been cited as one of the problems that precludes ultra-high field strength MRI from becoming a clinical reality. Here, we have developed an electromechanical solution. A 16-channel transmit-only TEM array [6] was designed and fitted with piezoelectric actuators mechanically linked to sliding cylindrical tuning and matching capacitors. The actuators were used to automate the tuning and matching process, based on the coil’s reflection coefficient [7,8]. Additionally a 32-channel receive-only loop [9] array was used in conjunction with the transmit-only array. These sets of arrays were constructed and tested for cardiovascular applications at 7 tesla [10].

II. THEORY

One of the most common transmitters at ultra-high field strengths is the TEM coil [9]. Figure 1 shows the most common implantation of the TEM coil: a microstrip resonator. This is a capacitively shunted λ/2 open-circuit resonator. While the use of true λ/2 microstrip resonators has been shown [11], their long lengths (~2.35 m at 1.5 T or 0.5 m at 7 T with an ε r=1) are impractical to implement. The coil is tuned to the Larmor frequency via two capacitors; one capacitor is usually a lumped-element fixed capacitor, C f, and the second is a variable “tuning” capacitor, C t. The resonator is matched to the transmission line using a series, variable “matching” capacitor, C m.

Since biological loads are not constant, they couple differently to the coil. Different loads will change the resonant frequency, quality factor and input impedance of each coil in the array. Therefore, it is requisite to re-tune and re-match each individual coil before imaging can begin. This is a tedious and time-consuming process. And unfortunately traditional methods of automating the tuning and matching process (e.g. varactors or digitally programmable capacitors)
do not have a power rating (up to 8 kW) that will withstand a standard MRI scan.

Figure 2 shows a design for a piezoelectric driven variable capacitor. In this design (a) is a non-magnetic micro-switch used to determine the “home” position of the capacitor (~0.5 pF); (b) is the piezoelectric actuator - SQUIGGLE SQL series actuator (New Scale Technology, Victor, NY); (c) is the actuator’s drive shaft, this shaft moves in and out; (d) the coupling shaft coupling the actuator (b) to the cylindrical capacitor (g); (e) is a compression spring supplying force to against the actuator; (f) is a front stop micro-switch used to determine the maximum position of the capacitor (~10 pF); (g) the cut-away view of the cylindrical capacitor showing the inner, sliding plate (h) and the outer stationary cylinder (i). The two capacitor plates are separated by a PTFE dielectric.

Figure 3 shows how two piezoelectric driven variable capacitors can be implemented on the standard TEM coil. Labels b-g on Fig. 3 are identical to Fig. 2 above; the front and back switches are not shown here.

Figure 4 shows the basic setup for the feedback driven automated tuning and matching process using piezoelectric actuation for a single coil. From left to right, the signal generator is a 1 kW pulsed power amplifier. The directional coupler is a -35 dB directional coupler fitted to the coil head. The reflectometer, also at the coil head, measures the magnitude and phase difference of the coupled forward and reflected signals and outputs the magnitude, $|\Gamma|$, and phase angle, $\angle \theta$, of the reflection coefficient (as DC voltages). In this case, the reflectometer is an AD8302 IC (Analog Devices, Norwood, MA). The piezo controller (E-861, Physik Instrumente (PI), Germany) generates the drive waveforms necessary to motivate piezoelectric actuation. Since only one piezo controller was used, opto-isolators were requisite to multiplex the drive waveforms to the correct actuator. Piezo $C_t$ and piezo $C_m$ are the tuning and matching capacitors, respectively. They are composed of the piezoelectric actuators mechanically linked to sliding cylindrical capacitors. The microcontroller ($\mu$C) samples and digitizes the reflection coefficient from the reflectometer and provides the digital logic for controlling the piezo controller and opto-isolators. The RF coil is the transmit-only TEM resonator.

Figure 5 shows the admittance chart and explains the empirical algorithm used for preliminary tests to automate tuning and matching. Here the blue arrows represent a change in the tuning capacitance and the red arrows represent a change in the matching capacitor. Both capacitors start at the home position (~0.5 pF). The matching capacitance is increased until the susceptance is 0.2+j B. Tuning capacitance can then be increased until the coil is tuned to the resonant frequency and matched to the transmission line. Due to losses in capacitors, often the matching capacitor needs to be adjusted during tuning, therefore impedance thresholds can be set (represented by the green lines). The impedance thresholds would allow the matching and tuning capacitors to be adjusted consecutively while keeping the conductance to $0.2 \pm 0.05$ siemens while reducing the coil’s admittance.
III. EXPERIMENTAL METHODS

A set of arrays were constructed and tested for cardiovascular applications at 7 tesla with automated tuning and matching capabilities using piezoelectric driven variable capacitors. Figure 5 shows one of these arrays, opened, to show each array consisting of an 8-channel transmit-only TEM array and a 16-channel receive-only loop array. For the transmit-only TEM array, the individual coil structures were 153 mm long with a 12.7 mm wide inner conductor and a 50 mm wide outer conductor, separated by a 19 mm thick PTFE dielectric bar with a low loss tangent and a relative permittivity ($\varepsilon_r$) of 2.08. The receive-only loop array is comprised of three rows of coils, the first and third row both contain five coils while the middle, or second row contains 6 coils. The overall length and width of the array sets are 33 and 43 cm respectively. Each individual coil is comprised of Number 10 AWG copper wire formed to a 9 cm circle, soldered to a RT/duroid 5880 (Rogers Corp, Chandler, AZ). Nearest neighbor coils had 2 cm overlap.

Every coil was independently tuned and matched to the Larmor frequency at 7T and decoupled from neighboring coils (capacitive decoupling for the TEM arrays; geometric and low impedance preamp decoupling for the loop arrays). Additionally every coil used PIN diode networks for active detuning.

A greedy, pseudo-gradient descent algorithm, based on the reflection coefficient, was developed to automate the tuning and matching process. This algorithm was used instead of the empirical solution shown above because it converged to a solution faster. In short, $C_m$ and $C_t$ were moved back to a home position (~0.5 pF). Then, using a turn-based method, the tuning and matching capacitance were varied to decrease the reflection coefficient. As the coil came closer to resonance, the capacitive step size decreased, thus, increasing the tune and match resolution.

IV. RESULTS

Prior to imaging, the array was loaded to a human torso and set on the MRI patient table. A 2 ms hard pulse (30 dBm at the coil head, 2% duty cycle) was used to tune and match each transmit coil. The effectiveness of the tune and match was measured, while the patient was still on the patient table by using a calibrated non-magnetic network analyzer.

On average it took 50 s to tune and match each coil and on average each coil had an $S_{11}$ greater than 20 dB with no coil reflecting more than 2.5% incident power.

Following tuning and matching, scout (GRE) images of the human thorax were acquired and $B_1^+$ shimming [4,5] was performed. Following this, T-GRAPPA images were acquired to access the array’s imaging performance. Fig. 7 shows two out of the 200 slices acquired; each image was acquired in 99 ms with 2.6 x 2.6 x 5.0 mm image resolution, and a parallel imaging reduction factor of 4 in the left-right direction. Due to the speed of the imaging sequence, breath holds were not needed. In both images, the heart is at end diastole, however,
Figure 7a was acquired during inspiration while Fig. 7b was acquired during expiration; this is most noticeable by the change in the position of the dome of the liver.

V. CONCLUSIONS

This array has shown the ability to automate the tuning and matching procedure with high fidelity; we believe this will become a significant paradigm shift in array design; benefitting both clinical and research studies.

VI. ACKNOWLEDGMENTS

This work was supported in part by NIH Grants P41 EB015894, R01 EB006835, 1R41EB013543-01 and R01 EB007327.

REFERENCES


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