Published in final edited form as:

Magn Reson Med. 2013 April ; 69(4): 1180–1185. doi:10.1002/mrm.24612.

Multi-turn transmit coil to increase B_1 efficiency in current source amplification

N. Gudino¹ and M.A. Griswold^{1,2}

¹Department of Biomedical Engineering, Case Western Reserve University, Cleveland, Ohio, United States

²Department of Radiology, University Hospitals of Cleveland, Cleveland, Ohio, United States

Abstract

Purpose—A multi-turn transmit surface coil design was presented to improve B_1 efficiency when used with current source amplification.

Methods—Three different coil designs driven by an on-coil current-mode class-D (CMCD) amplifier with current envelope feedback were tested on the benchtop and through imaging in a 1.5 T scanner. Case temperature of the power field-effect transistor (FET) at the amplifier output stage was measured to evaluate heat dissipation for the different current levels and coil configurations. In addition, a lower power rated device was tested to exploit the potential gain in B_1 obtained with the multi-turn coil.

Results—As shown both on the benchtop and in a 1.5 T scanner, B_1 was increased by almost three-fold without increasing heat dissipation on the power device at the amplifier's output using a multi-turn surface coil. Similar gain was obtained when connecting a lower power rated FET to the multi-turn coil.

Conclusion—In addition to reduce heat dissipation per B_1 in the device, higher B_1 per current efficiency allows the use of FETs with lower current ratings and lower port capacitances which could improve the overall performance of the on-coil current source transmit system.

Keywords

coil array; MRI transmission; transmit efficiency; current source amplification

Introduction

Array coils have clearly revolutionized much of modern MRI practice, and new research is constantly pushing the limits of this technology. Research efforts in this area have focused on receiver arrays [1], with an ever increasing interest in transmit arrays [2,3]. The majority of advances in array coils have come through increases in the sheer number of array elements, which are currently on the order of 128 elements for receiver coils [4,5] and 64 elements for transmitters [6]. While much of the receiver array problem has been reduced to implementation issues, there remain significant uncertainties about the configuration of an optimal transmit array, amplifier setup and control system. Currently our preferred technology for the implementation of transmit arrays is the direct control of B_1 field achieved by current source amplification [7-9]. This method has the advantages of load insensitivity, and isolation in a multiple-transmit setup through the amplifier decoupling

method. Recently we presented an on-coil current-mode class-D (CMCD) with envelope feedback amplifier as an alternative design for the implementation of parallel transmitters [9]. With the amplifier loaded with a $7\ \Omega$ impedance we measured output power around 300 W RMS and peak current above 10 A. These power and current ratings required the use of high current RF power metal-oxide semiconductor field-effect transistors (MOSFETs) at the CMCD output stage.

High power RF FETs such as this have non-negligible voltage-modulated capacitances that increase with nominal power and have an effect on the amplifier performance. Drain-source capacitance, for example, had a direct effect on the amplifier decoupling as well as on the phase of the RF carrier [9]. In general, a device with lower capacitance values across the whole range of operating drain voltages would result in improved overall system performance but these devices normally come with reductions in the nominal drain current. Since the target load is usually a relatively fixed value, the exact choice of device is generally the only available variable in this optimization problem.

Here we propose a rather simple change to each element in a current-mode amplifier configuration that could allow other improved choices for current source amplification. Multiple-turn surface coils are not used in practice, since they generally offer reduced performance when used in a receive mode at clinical field strengths. However, in this work we propose a transmit setup where the on-coil CMCD amplifier with envelope feedback drives a multi-turn transmit surface loop. Since the current-mode devices directly drive current into the load and not voltage across a load, we show that we can exploit the coil geometry to increase the amplitude of the transmit B_1 . Both benchtop and scanner tests proved a gain in B_1 when increasing number of turns at same amplifier output current and similar heat dissipation without any modification to the amplifier unit. This coil design together with the presented amplifier topology should help to the implementation of optimized multiple-channel transmit systems.

Materials and Methods

Three different coils were designed: a single-turn (coil A), two-turn coil (coil B) both printed on one-layer 1.6 mm thick FR-4 board and four-turn coil (coil C) printed on two-layer 1.6 mm thick FR-4 board (Fig. 1). For all coils the trace width was 2.5 mm, the space between turns was 1.4 mm and the outer dimensions were 90 mm $\times$ 90 mm. We set three capacitors splitting per turn to resonate each coil at 63.66 MHz when loaded with a 2000 ml saline phantom. The input impedance for the three different coils (connected to a 25 cm cable with a 84 pF capacitor in series to resonate the cable and coil combined structure at 63.66 MHz at the amplifier terminals) was measured through a network analyzer (Agilent Technologies, E5061A) in unloaded and loaded condition. Since the presented coil configurations are intended for implementation in a multi-element array, we placed a pair of each of the coil designs modified with a 50 Ω matching network and measured the coupling between elements using the network analyzer. With the coils loaded with a large cylindrical phantom (7300 ml), S21 was measured at 4 cm and 2 cm separations and compared among the different configurations.

We connected each of the coils to the same CMCD amplifier with envelope current feedback as shown in Fig. 2. As detailed in a previous work [9], this amplifier has three subsystems: 1) a digital to analog interface and preamplifier that takes as input a digital encoded RF carrier and generates high enough voltage and current to switch the active devices on the output stages 2) the CMCD amplifier that has two power MOSFETs switched in push-pull configuration at the RF carrier's frequency and 3) the amplitude modulation system that drives the supply voltage to the CMCD stage to track the RF envelope target

signal through a closed-loop feedback with a current sensor located at the amplifier's output. In this testing setup, the variable that sets the amplitude of the output current (digital encoded RF envelope $\Sigma\Delta$) was kept at the same level for all the coils. The input current to the CMCD stage (I_{DD}) for a given $\Sigma\Delta$ value was estimated from a differential voltage reading across a $0.1\ \Omega$ (1% tolerance) shunt. Note that for the CMCD topology ideally the input current is split in half due to the bridge configuration and the harmonic currents are driven through an output resonant filter (formed by the MOSFET's output capacitance in parallel with an external inductor). Thus at the RF bandwidth the input current behaves as a square signal and the current on the coil can be approximated as

$$\widehat{I}_{Coil} \sim \left(\frac{4}{\pi}\right) \frac{I_{DD}}{2} \quad [1]$$

From previous power and efficiency measurements the error that resulted from this current estimation was below 15 %. Although we were mainly interested in the gain in B_1 per amplifier input current we used Eq. 1 to estimate the current driven by the coil when needed.

The observed output B_1 field generated in the center of the coils was estimated using a 2 cm diameter pick-up loop coupled to the coil and connected through a 20 dB attenuator to an oscilloscope (Agilent DS07104B). The peak-to-peak voltage (V_{PP}) generated in the center of the coils was converted into B_1 field values using a conversion factor of $6.4\ \mu\text{T}/V_{PP}$ as estimated based on V_{PP} vs. I_{Coil} curves, and coil geometry. Mean and standard deviation were calculated from three replicates in all voltage measurements.

We exploited the load insensitivity of the current source transmitter and tested two coil-phantom configurations. A first setup where all coils were set at same distance from the phantom through a 11 mm spacer and a second setup, where this distance was changed such that the loading to the amplifier from coil B was similar to coil A while the loading from coil C was made similar to coil B. In addition, since heat dissipation is an important factor for the on-coil amplifier, the temperature on the MOSFET's case (critical hot spot) was measured with a fiber optic temperature sensor (OpSens L804-0083-05, -40 to 125 °C, Canada) after 5 minutes of continuous operation. Following the same measurement methods, the gain in B_1 was compared between coil A and coil B when driving 5 A and 10 A approximately.

In addition, the gain obtained with the multi-turn coil design was compared to the gain obtained through current transformation at the amplifier's output. Two 2:1 air-core trifilar transformers, commonly used in RF applications (AN20-002. Mini-circuits) [10], were built with a 0.8 mm diameter copper magnet wire wound on circular plastic formers. The first transformer (TA) had 5 turns wound over a former with 0.5 cm internal diameter, 1.2 cm external diameter and 2 mm thickness. The second transformer (TB) had 3.5 turns over a former of 0.4 cm internal diameter, 0.95 cm external diameter and 1 mm thickness. Using a 50 Ω setup we estimated the transformers insertion losses by connecting the high impedance winding to port 2 of the network analyzer and the low impedance was connected to port 1 through an impedance matching network after resonating any primary leakage inductance with a series capacitor. Insertion losses were estimated as the difference of S_{21} measurements in this setup and the corresponding insertion losses of the matching pad. In the amplifier implementation, to have an ideal 1:2 current ratio, each transformer was connected with its high impedance winding to the amplifier's output and its low impedance side to coil A. Similarly to the previous experiments, drain current was kept constant such that approximately 5 A or 10A was driven into the single-turn coil according to Eq. 1, and V_{PP} was measured at the center of the coil for the three different setups.

To test the potential to use different devices that can better exploit the higher gain in B_1 , we replaced the setup with coil A driven by the MRF275G (M/A-COM semiconductor) by a setup with coil C driven by the MRF6V2010NR1 (Freescale). For comparison, we obtained the input current values that generated similar B_1 values than those previously acquired with the high power device. We also measured the output impedance of the amplifier in each of these configurations. As performed in [9], for the MRF275G the resonant output filter was formed by an external inductor that resonates the output capacitance of the MOSFET when biased at 15 V. In the case of the MRF6V2010NR1, the filter was formed by a 51 μH external inductor and a 100 pF external capacitor in order to compensate for the low output capacitance of this device (< 30 pF).

Finally, we tested the different coils in a 1.5 T scanner (Espree, Siemens, Germany), by connecting each of them to the same (high power MRF275G) amplifier unit through a BNC connector. The current-mode amplifier was used for transmission and the body coil was used for signal reception. Images for each design were acquired while keeping the same input value of RF envelope, thus similar amplifier output current. Flip angle maps were obtained through the double angle method (DAM) [11] using a GRE sequence with 2 ms RF pulse, 10 ms TE, 3 s TR, 20 cm $\times$ 20 cm FOV. Average flip angle and B_1 values were calculated at same 5-pixel ROI for each of the transmit coils.

Results

Input impedances for the different coil designs are shown in Table 1. Higher unloaded impedance was measured as expected due to the increased length of copper traces in the multi-turn design. In addition, in the loading condition the effective impedance was higher due to increased coil-sample coupling.

In the coupling test setup, S21 values were between -6 dB and -7 dB approximately for all pair of coils located at the same plane and separated 2 cm and 4 cm of distance. This confirmed that the multi-turn elements did not present higher intrinsic coupling values than their single-turn counterparts.

Independent of the loading conditions, the current setting for B_1 and temperature measurements was such that peak current on the coil was close to 5 A or 10 A (Table 2). The resulting B_1 at the coil center is shown in Fig. 3 (a,b), and temperature increment (ΔT) on the power MOSFETs is shown in Fig. 3 (c,d) when driving approximately 5 A to all coils. At this current level, and with all coils at 11 mm distance from the phantom, B_1 was 1.7 higher for coil B than coil A, and 1.5 higher for coil C than coil B, resulting in a total gain in B_1 of approximately 2.6 from coil A to C [Fig. 3(a)]. Figure 3(c) shows that higher B_1 was possible without increasing heat dissipation. Figure 3 (b,d) shows the results when changing the coil-phantom distance such that the loading to the amplifier from coil B is similar to coil A and from coil C to coil B. The gain in B_1 was similar and approximately 1.9 higher for coil B than coil A and 1.5 higher from coil C than coil B, resulting in a total gain of 2.85 for this case [Fig. 3(b)]. Figure 3(d) confirmed that as in the previous setup heat dissipation was not significantly increased at the higher B_1 amplitude. Figure 4 shows the result of driving coil A and coil B at two different current levels. By driving coil B at 5 A we obtained a similar B_1 as compared to driving coil A at approximately 10 A [Fig 4(a)], while at the same time, ΔT changed from 9.6 $^{\circ}\text{C}$ to 4.6 $^{\circ}\text{C}$ indicating an important reduction in heat dissipation [Fig. 4(b)].

Based on S21 measurements performed with the 50 Ω setup, transformer insertion losses at 63.66 MHz were -0.5 dB and -1.6 dB for TA and TB respectively. Figure 5 shows the gain

in B₁ when connecting the 2:1 trifilar transformers. A gain in B₁ of 1.36 was achieved with TA; this was 20 % lower than the gain obtained with the two-turn coil.

Similar B₁ values were possible at a lower current range for the lower power rated device connected to coil C [Fig. 6(a)]. For this setup, a B₁ up to 46 μ T at I_{DD} of approximately 1.5 A was possible. This represented approximately three-fold reduction in current for same B₁ value when compared with the MRF275G. Figure 6(b,c) shows amplifier output impedance versus drain voltage for each setup.

Figure 7(a) shows GRE images of the 2000 ml saline phantom when transmitting RF with the different transmit coils and receiving signal with the body coil of the scanner. Note that a signal inversion band was possible for coil C at a similar current level as applied to coils A and B. Figure 7 (b) shows average flip angle and B₁ amplitude values at the selected 5-pixel ROI. When compared to benchtop results, a similar gain in B₁ was obtained from coil B to coil C and slightly higher gain was obtained from coil A to B. This small difference in observed performance is most likely the result of a shift in coil A's position relative to phantom and/or image plane in addition to suboptimal DAM map calculations resulting from the low flip angle (<50°) in the selected ROI. Regardless of this difference, images confirmed the gain in B₁ when driving the different coils with more turns for any given amplifier.

Discussion

Using a multi-turn transmit coils we have shown that the effective B₁ of a current-mode transmit system can be increased, even though these coils present a higher load at the amplifier output. With the multiple-turn design we were able to generate about 3 times the B₁ value obtained with a single turn without increasing output current or dissipation on the power FET. This higher efficiency can allow either significantly increased effective power per amplifier, or can allow the use of lower current/power rated FETs with reduced port capacitances, which should improve the overall performance and cost of the on-coil current-mode amplifier. As shown in the Results Section, the lower output capacitance of the MRF6V2010NR1 provided larger impedance over a larger range of V_{DD}. This effect should offer advantages in terms of phase modulation and amplifier decoupling. Using this nominally lower power rated FET in combination with the configuration in coil C would clearly be able to deliver sufficient B₁ field for MR imaging even though the nominal power rating of this device is only around 10 W RMS in the 10 MHz to 450 MHz range.

Higher B₁/I values will reduce the current requirements at a given nominal drain voltage simplifying design specifications and increasing amplifier performance. In addition, we showed that increasing B₁/I efficiency through the multi-turn coil design can minimize heat dissipation on the RF power FET, which is an important advantage for truly on-coil amplifier implementations. This is expected for this amplifier topology [12], since heat dissipation on the power device is dominated by $(I_{DD})^2 R_{DS_ON}$, where R_{DS_ON} is the on-state resistance of the power MOSFET. Thus for the current source operation its value should be similar for all load conditions at a given envelope shape, amplitude and duty cycle. In general, lower drain current specification for the semiconductor device will also keep R_{DS_ON} low and closer to its rated value which should improve power efficiency across the operating current range.

We were unable to reproduce the gain in B₁ of the two-turn coil setup with the simple 2:1 RF trifilar transformers presented here. Based on the conventional 50 Ω low-power measurement setup, TA had low insertion losses and comparable to commercial devices but when implemented on the amplifier's high-power setup higher losses became evident from

the lower measured current ratio. This is expected to be worse at higher transformation ratio where copper losses would be increased by higher secondary current values. While it is likely that transformers with slightly higher performance could be constructed, we invested a significant amount of effort in this test, since a transformer solution would be advantageous due to its ease of integration into current designs. However, none of the transformer configurations tested was comparable to the multi-turn coils in either ease of construction or efficiency.

By moving to a coil with more turns, we do increase the potential to encounter a voltage limit in the output stage because the effective load impedance is increased. However, the switch-mode structure of the CMCD is such that each of the FETs is alternatively switched between a high voltage regime and a high current regime ideally with no overlap between these states. Thus the increase in the voltage seen by the FET does not directly translate into an increase in power dissipated in the device, since ideally no current flows through the FET in this state. This was confirmed by the heating experiments shown here. The increase in load impedance had only minimal to no effect on the device heating.

Thus we believe that the use of multi-turn coils in combination with current-mode amplifiers is a promising approach to the design of parallel transmit systems. The number and configuration of the turns provides additional degrees of freedom that effectively can be used to increase the B_1 efficiency of an amplifier. There may be other degrees of freedom, such as field shaping through differential design of the multiple loops that may be possible and will be the subject of future work in this area.

Acknowledgments

Research sponsored by Siemens Healthcare. The authors would like to thank the research groups at NHLBI and NINDS (National Institute of Health, Bethesda) for their support.

References

1. Roemer PB, Edelstein WA, Hayes CE, Souza SP, Mueller OM. The NMR phased array. *Magn Reson Med*. 1990; 16(2):192–225. [PubMed: 2266841]
2. Katscher U, Börnert P, Leussler C, van den Brink JS. Transmit SENSE. *Magn Reson Med*. 2003; 49(1):144–150. [PubMed: 12509830]
3. Zhu Y. Parallel excitation with an array of transmit coils. *Magn Reson Med*. 2004; 51(4):775–784. [PubMed: 15065251]
4. Hardy CJ, Giaquinto RO, Piel JE, et al. 128-channel body MRI with a flexible high-density receiver-coil array. *J of Magn Reson Imag*. 2008; 28(5):1219–1225.
5. Schmitt M, Potthast A, Sosnovik DE, Polimeni JR, Wiggins GC, Triantafyllou C, Wald. A 128-Channel Receive-Only Cardiac Coil for Highly Accelerated Cardiac MRI at 3 Tesla. *Magn Reson Med*. 2008; 59(6):1431–1439. [PubMed: 18506789]
6. Feng K, Hollingsworth N, McDougall M, Wright S. A 64-Channel Transmitter for Investigating Parallel Transmit MRI. *IEEE Trans Biomed Eng*. 2012; 59(8):2152–2160. [PubMed: 22552545]
7. Kurpad KN, Wright SM, Boskamp EB. RF current element design for independent control of current amplitude and phase in transmit phased arrays. *Concepts in Magn Reson Part B: Magn Reson Eng*. 2006; 29B(2):75–83.
8. Lee W, Boskamp E, Grist T, Kurpad K. Radiofrequency current source (RFCS) drive and decoupling technique for parallel transmit arrays using a high-power metal oxide semiconductor field-effect transistor (MOSFET). *Magn Reson Med*. 2009; 62(1):218–228. [PubMed: 19353658]
9. Gudino N, Heilman JA, Riffe MJ, Heid O, Vester M, Griswold MA. On-coil multiple channel transmit system based on class-D amplification and pre-amplification with current amplitude feedback. *Magn Reson Med*. 2012; 1002/mrm.24462
10. Carr, J. *J RF Components and Circuits*. Oxford; Newnes: 2002.

11. Insko EK, Bolinger L. Mapping of the Radiofrequency Field. *J Magn Reson, Series A*. 1993; 103(1):82–85.
12. Kobayashi H, Hinrichs JM, Asbeck PM. Current-Mode Class-D Power Amplifiers for High-Efficiency RF Applications. *IEEE Trans on Microwave Theory and Techniques*. 2001; 49(12)

Figure 1. Layout of the different coils. Coil A with one turn, coil B with same dimensions and two turns and coil C built as two layer coil of same dimension than coil B and four turns.

Figure 2.

Setup to measure the effect in B_1 gain of the different coils. Current was set by the control system and is independent of coil-load condition. Current input to the CMCD stage (I_{DD}) was set such that initially current in the coil was approximately 5 A peak for all load conditions.

Figure 3.

B1 field generated in the center of each coil at approximately 5 A output current with coil-phantom distance set to 11 mm for all coils (a), and with this distance modified such that coil B had similar loading than coil A ($d=43$ mm) and coil C had similar loading than coil B ($d=36.5$ mm) (b). Temperature increment was measured on the MOSFET's case after 5-minute continuous operation at 2% duty cycle to evaluate heat dissipation from the electronics in both setups (c, d).

Figure 4. B_1 at the center of coil A and coil B at two different current levels ($I \sim 5$ A and $2I \sim 10$ A) (a), and temperature increment on the MOSFET's case (b).

Figure 5.

B_1 measured at the center of coil A without a transformer (NT) and with two types of 2:1 trifilar transformers connected at the amplifier's output (TA and TB). I_{DD} was set close to 8 A for all measurements to compare the gain of the two different transformers to the single-turn setup.

Figure 6. B1 versus input current for low power MOSFET (10 W, MRF6V2010N, Freescale) connected to 4-turn coil (coil C) and high power MOSFET (250 W, MRF275G, Maacom) connected to single-turn coil (coil A) (a). Output resistance (b) and reactance (c) versus V_{DD} for both FETs.

Figure 7. GRE images obtained with the three different transmit coils (coil A, B, C from left to right) at same input envelope amplitude value and 2 ms pulse length set by the system's controller (a). Average flip angle and B1 values calculated over a 5-pixel region centered at the highlighted pixel on maps obtained through DAM.

Table 1
Input impedance of the different transmit coils connected to a 25 cm coaxial cable and resonated in the loaded condition at 63.66 MHz

Coils	$Z_{\text{UNLOADED}} [\Omega]$	$Z_{\text{LOADED}} [\Omega]^*$
A	$0.9 - j 0.4$	$1.7 - j 0.2$
B	$1.2 - j 0.5$	$4.1 - j 0.2$
C	$2.2 + j 1.3$	$11.7 - j 0.3$

* Loaded with 2000 ml saline through 11 mm spacer.

Table 2
Current setting for the on-coil current-mode amplifier driving the different coils at different loading conditions

Spacer (mm)	Coils	Z [Ω]	$\Sigma\Delta$	I _{DD} [A]	I _{Peak} ^{**} [A]
11	A	Z _A	0.15	8.35±0.18	5.32±0.12
11	B	Z _B	0.15	8.08±0.24	5.14±0.15
11	C	Z _C	0.15	8.05±0.46	5.12±0.29
11	A	Z _A	0.3	16.40±0.26	10.44±0.17
11	B	Z _B	0.3	16.73±0.85	10.65±0.54
43	B	Z _A	0.15	8.01±0.20	5.10±0.13
36.5	C	Z _B	0.15	8.2±0.17	5.22±0.11

* estimated peak current on the coil from IDD measurements.