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**An 8-element Tx/Rx Array Utilizing MEMS Detuning Combined with 6 Rx Loops for $^{19}$F and $^{1}$H Lung Imaging at 1.5 T**

Adam Maunder[1], Madhwesha Rao[1], Fraser Robb[1,2], Jim M. Wild[1*]

1. POLARIS, Imaging Group, Department of IICD, University of Sheffield, UK.
2.GE Healthcare, Aurora, OH, USA.

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***Corresponding author:**

Jim M. Wild
Polaris, Imaging Group, Department of Infection, Immunity & Cardiovascular Disease
University of Sheffield
C Floor
Royal Hallamshire Hospital
Sheffield
S10 2JF
UK

Telephone: + 44 (0)114 2159141
Email: j.m.wild@sheffield.ac.uk

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## Abstract

**Purpose:** To firstly improve the attainable image signal-to-noise ratio (SNR) of $^{19}F$ and $^{1}H$ $C_3F_8$ lung imaging at 1.5T using an 8-element transmit/receive (Tx/Rx) flexible vest array combined with a 6-element Rx-only array, and secondly, to evaluate MEMS for switching the array elements between the two resonant frequencies.

**Methods:** The Tx efficiency and homogeneity of the 8-element array was measured and simulated for $^{1}H$ imaging in a cylindrical phantom and then evaluated for in-vivo $^{19}F/^{1}H$ imaging. The added improvement provided by the 6-element Rx-only array was quantified through simulation and measurement and was compared to the ultimate SNR (uSNR). It was verified through the measurement of isolation, that MEMS switches provided broadband isolation of Tx/Rx circuitry such that the $^{19}F$ tuned Tx/Rx array could be effectively used for both $^{19}F$ and $^{1}H$ nuclei.

**Results:** For $^{1}H$ imaging, the measured Tx efficiency/homogeneity (mean$\pm$percent standard deviation; $6.79\mu T/\sqrt{kW} \pm 26\%$) was comparable to that simulated ($7.57\mu T/\sqrt{kW} \pm 20\%$). The 6 additional Rx-only loops increased the mean Rx sensitivity when compared to the 8-element array, by a factor of 1.41x and 1.45x in simulation and measurement, respectively. In regions central to the thorax the simulated SNR of the 14-element array achieves >=70% of the uSNR when including noise from the matching circuits and preamplifiers. A measured MEMS switching speed of 12 μs and added minimum 22 dB of isolation between Tx and Rx was sufficient for Tx/Rx switching in this application.

**Conclusion:** The described single tuned array driven at $^{19}F$ and $^{1}H$, utilizing MEMS technology, provides excellent results for $^{19}F$ and $^{1}H$ dual-nuclear lung ventilation imaging.

**Keywords: Transceive Array, fluorine-19 MRI, Microelectromechanical Systems (MEMS), PIN diode, lung MRI,**

## Introduction

For the clinical diagnosis and monitoring of lung diseases such as late-stage cystic fibrosis, where morphological and physiological changes may be significant, current state-of-art proton MR imaging of the lungs is clinically useful. However, in lung diseases such as early-stage cystic fibrosis or emphysema structural changes are less obvious and the low tissue density of the lungs makes imaging a challenge (1). Hyperpolarized (HP) gas MRI has the capacity to image lung microstructure and function with high signal-to-noise (2), which enables detection of impaired function in diseased lungs with high sensitivity. Hyperpolarized gas imaging (with $^{3}He$ and $^{129}Xe$) has become a well-established method for ventilation imaging, with numerous studies demonstrating its sensitivity to early regional signs of obstructive airways disease and its correlation with common global measures of lung function (3, 4).

### Fluorinated Gas Imaging

MRI of inert fluorinated gases is an emerging method for pulmonary ventilation imaging, but image quality is constrained by low spin density (~2 $\times 10^{19}$ atoms/cm$^3$ for gases (5) vs. ~$7\times10^{22}$ atoms/cm$^3$ for $^{1}H$) and the short $T_2^*$ (< 3 ms (6)). Therefore, in-vivo image resolution attainable within a single breath-hold is limited. Currently, fluorinated gas lung imaging is not as well characterized as HP gas imaging in clinical studies, due to the greater research interest in HP gas because of the intrinsically higher attainable image quality. Furthermore, the specific physical properties of the noble gases influence their scope for clinical translation, for instance, $^{3}He$ is more diffuse (7), while $^{129}Xe$ dissolves into the blood-stream and can be used to characterize gas exchange (8). Recently, there have been significant improvements in sequence optimization for fluorinated gas imaging using ultrashort echo time and steady state free precession methods (9). However, to date, there has been no clear demonstration that fluorinated gas imaging can be used routinely to provide robust quantitative measures of lung function. Therefore, to demonstrate that ventilation imaging can be used in a similar manner in a clinical setting, it is necessary to ensure that the highest possible SNR is obtained. In this study, the potential increase in SNR that can be achieved in $^{19}F$ lung MRI by improved radio-frequency (RF) coil design was explored.

Most commonly, $^{19}F$ lung imaging has been performed with the use of single-channel transmit/receive (Tx/Rx) coils (10, 11). Receive arrays provide improved SNR, while maintaining a large field of view (FOV), by combining individually received signals from multiple isolated smaller RF coils (12). Therefore, the use of a Rx array for $^{19}$F lung imaging, as in reference (13), is potentially beneficial in the acquisition of higher resolution ventilation images. Additionally, an emerging method of investigating lung function relates the change in $^{1}$H signal with different inflation volumes (14), based on dynamic Fourier decomposition methods (15) with registered images (16). Therefore, the improvement of SNR from lung parenchyma in $^{1}$H imaging with the use of an array for both $^{19}$F and $^{1}$H nuclei could be beneficial for the combined investigation of complementary structural and functional methods of lung imaging.

**The Choice of Coil Topology For Tx/Rx**

The commonly used combination of a birdcage Tx coil and dedicated Rx array offers a solution for increased SNR but has two potential disadvantages for the application of $^{19}$F lung imaging. First, the patients participating in ventilation imaging often have debilitating conditions such as COPD or asthma, with hyper-inflated chest volumes, so coil volume, patient comfort and minimizing time spent in the MRI bore are high priorities. A body-sized rigid birdcage coil within the scanner bore is cumbersome because it reduces available space for a nested Rx array, increases set-up/removal times and it also restricts space and movement in the bore, which affects patient comfort (for typical ventilation imaging it is only necessary that the volunteer/patient remains motionless during breath-hold). Secondly, from an RF engineering perspective, a large $^{19}$F birdcage coil will couple strongly to the integrated system $^{1}$H body coil and any $^{1}$H/$^{19}$F Rx arrays.

Conventional strategies of detuning Tx coils from Rx coils employ switches, such as PIN diodes in-line with the Tx coil, to present a high impedance during Rx and the use of LC traps in-line with forward biased PIN diodes during Tx to isolate Rx coils. To decouple $^{1}$H Tx/Rx coils from multinuclear Tx/Rx coils, passive LC traps are typically introduced that are resonant at the alternate frequencies. However, for $^{19}$F imaging, the resonant frequency is only 6% lower than the $^{1}$H frequency, so a large loss would be introduced into both coils by using traps that would greatly reduce Tx efficiency and Rx sensitivity. Therefore, for $^{19}$F imaging different strategies must be employed, such as that in reference (17), where varactor capacitors were used on the Rx elements to tune the coils and adjust the preamplifier input and output matching between $^{1}$H and $^{19}$F imaging. A transceiver array that uses the same elements for $^{1}$H and $^{19}$F imaging would eliminate the need for complex and cumbersome strategies of detuning the different $^{1}$H and $^{19}$F arrays, as well as obviating the need for Tx/Rx detuning of separate $^{19}$F and $^{1}$H Tx coils.

For these reasons, and to improve subject comfort and provide a robust multi-element coil design that simplifies detuning, an 8-element Tx/Rx array was designed for both $^{1}$H (64 MHz) and $^{19}$F imaging (60 MHz) at 1.5 T. The 8-element Tx/Rx array was designed as a dual-row (anterior-right-posterior-left) flexible vest coil that conforms to the geometry of the adult thorax. Previously, it has been demonstrated that the same coil tuned for $^{19}$F imaging may also be used for $^{1}$H imaging without adjustment of tuning (18). The impact on performance with this strategy was investigated, with the expectation that Tx efficiency will be reduced as a consequence of the mismatch between the power amplifier and coil input impedance at the $^{1}$H frequency, as was found in reference (19).

**Switching Between $^{1}$H and $^{19}$F Frequencies and Tx/Rx Isolation**

In a Rx array, LC traps that are resonant at a single frequency are typically switched by PIN diodes to alternate between Tx and Rx states. However, for imaging both $^{19}$F and $^{1}$H a method that covers multiple frequencies must be used. Microelectromechanical systems (MEMS) have been proposed as a method for decoupling coils with a number of advantages (e.g. broadband operation, low DC power, fast switching (20-22) over the strategy of utilizing tanks circuits enabled by PIN diodes (23). The use of MEMS in MRI RF coil design is a relatively new development and has required improvements over previous MEMS switch designs for robust implementation. Namely: higher stand-off voltage, simultaneous high-beam reliability with high conductance, and non-magnetic design (24). In subsequent applications, the MEMS device has been employed for detuning in Rx arrays (23), coil dual-tuning (19), reconfigurable coil arrays (22), wireless power transfer in the MR system (24) and coils for $B_0$ shimming (25). Therefore, to protect the Rx chain from coupling to Tx power, MEMS switches were incorporated in the Rx detuning network. The broadband nature of MEMS means that the coil can also be used for $^{1}$H imaging without any additional circuitry. With this method, the issue of coupling between Tx and Rx circuitry is mitigated for both $^{19}$F and $^{1}$H frequencies.

**Improving Sensitivity with Additional Rx Elements**

The proposed 8-element MEMS-switched Tx/Rx array was designed for reasonable Tx homogeneity, and thus has limited Rx sensitivity in central regions of the lungs. Previously, the use of a Tx/Rx array with additional Rx-only elements has been predominantly employed at ultra-high field strengths ($\geq$4.7 T) (26), (27). It has been demonstrated that although Tx/Rx arrays designed for good Tx homogeneity may not be optimal in Rx configuration, the addition of Rx-only elements targeting regions of low sensitivity may be an equally good alternative, with less complexity to designs that have totally separate Tx-only and Rx-only arrays.

Consequently, here a 6-element Rx-only array was added, requiring detuning from the 8-element Tx/Rx array during Tx for both $^{1}H$ and $^{19}F$ imaging.

**Assesment of SNR**

There are theoretical constraints on the maximum SNR obtainable based on electromagnetic principles, termed the ultimate SNR (uSNR) (28). Methods for calculating the uSNR have been developed for simple geometries such as a cylinder or box (29, 30), a sphere (31, 32) and even in realistic models of the human head (33). For the purpose of this study, the simulated performance of the designed 14-element Rx array was compared to the uSNR in a simple cylindrical phantom model corresponding to the size of an adult human torso with homogeneous electrical properties, to be related to any future $^{19}F/^{1}H$ lung designs.

Finally, 3D in-vivo fluorinated gas ventilation images and structural $^{1}H$ images were obtained with the coil ensemble, demonstrating the achievable image quality with an optimized image sequence (9) and the designed coil array.

## Methods

### Coil Array Layout

The 2D conductor layouts with labeled dimensions are shown for the 8-element Tx/Rx array and 6-element Rx-only array in Figure 1a. The use of dual-transceiver array row designs has been shown to provide good performance in Tx (34, 35) and simulations indicate similar specific absorption rate (SAR) and Tx efficiency metrics to those of equivalent single row designs (36). For the purpose of lung imaging with a flexible array, the use of more than 4-elements on a single row does not allow for easy access and movement of the arms. Multi-row designs are also routinely used to improve SNR along the longitudinal direction (37). Thus, a dual-row design was used here. The original prototype of the 8-element transceiver array (38) was found to have lower sensitivity in the center of the lungs indicating that improvements could be made. To address this a 6-element Rx array was designed with elements placed on the front and back, as in typical cardiac designs targeting the center of the chest (39). The 6-element Rx-only array was overlaid on top with three anterior elements and three posterior elements. The 8-element Tx/Rx array was constructed on a flexible printed circuit board (PCB), while the conductive elements of the 6-element Rx array consisted of copper tape placed on flexible Teflon sheet. The HFSS (ANSYS, Canonsburg, PA) simulation model and 3D model for the combined 8-element Tx/Rx array and 6-element Rx array is shown in Figure 1c.

**8-element Tx/Rx Coil Array Design**

The dimensions of the flexible 8-element Tx/Rx coil (PCB design schematic in Figure 1a) were selected to be suitable for the majority of adult body sizes. Left and right coils are wider (35.5 cm) than anterior and posterior coils (30.5 cm). In simulation, the 8-element Tx/Rx coil array was designed to conform to the geometry of a cylindrical ellipse with minor axis radius of 16.8 cm, major axis radius of 20.8 cm and 42 cm length (3D model shown in Figure 2).

The Tx power division, the Tx and Rx matching, and the Tx/Rx decoupling circuits for superior-inferior coil pairs are contained in the white boxes shown in Figure 1a and 3D models in Figure 2. The Tx power for the four coil pairs was split equally using lumped element hybrid couplers with 0°, 90°, 180°, and 270° phases. Coaxial cables run from power dividers that provide a 180° phase shift between anterior-posterior coils and left-right coils as in reference (24) (cables are shown schematically in Figure 1b). The -3dB 90° and 180° couplers were constructed according to standard circuits shown in reference (40). The scattering parameters of the power dividers were measured using an Agilent E5061A Network Analyzer (Keysights, Santa Clara, CA) and it was determined that the reflection coefficient was less than -15 dB at both 60 MHz and 64 MHz providing that all of the ports were matched and that the insertion loss was $\sim 0.5$ dB, with $3.5 \pm 0.1$ dB transmission to both output ports.

The decoupling between elements, the matching, the preamplifier decoupling, and Tx/Rx decoupling for the superior-inferior pairs of the 8-element Tx/Rx array are provided by the networks shown in the circuit diagram, and constructed circuits, in the Supporting Information Figure S1. The coils were tuned/matched to the $^{19}$F frequency when placed on a cylindrical phantom representing a human load (see Phantom Imaging section of methods for details). Adjacent coils were decoupled by critical overlap, while capacitive decoupling was used to decouple superior-inferior pairs (41). Matching was performed with as $\lambda/4$ T-line LC network (40). The T-line connected to the preamplifiers by $\lambda/2$ cables, provides preamplifier decoupling. Wilkinson power dividers (40) provide equal power division to superior-inferior coils during Tx, and in Rx are isolated from the coils.

**6-element Rx-only Coil Array Design**

The geometry of the 8-element Tx/Rx array in this study allows an efficient layout of cables and circuitry that does not impede subject access or comfort while providing satisfactory Tx homogeneity. However, in the direct center of the imaging region, and between coils, the Rx sensitivity was poor due to the double-row design, motivating the addition of the additional 6-

element Rx-only array. The diagram of the 6-element Rx array conductor layout (Figure 1a) shows each rectangular coil is 25 cm long and 13.5 cm wide. The 6-element Rx array could potentially provide a slightly higher sensitivity when placed directly on the body, but by affixing the array to the outside of the flexible 8-element array the comfort and space in the coil is maximized, while the additional < 0.5 cm closer proximity is small when compared to the >10 cm radius of an average adult human torso.

The decoupling between elements, Tx/Rx detuning, and matching and preamplifier decoupling for the 6-element Rx-only array are described as follows and shown in Figure 1c. Element decoupling: adjacent coils are decoupled by critical overlap. Tx/Rx detuning: series active detuning using LC traps and PIN-diode (MA4P7435F-1091T, MACOM, MA, USA) switching was used for both $^{1}$H and $^{19}$F frequencies. For each coil, there are two traps tuned to 64 MHz and two tuned to 60 MHz shown with inductors $L_H$ and $L_F$ in Figure 1c, respectively. A forward-biased diode placed at the input of the LC balun in Figure 1c will provide a high input impedance to be presented in line with the coil, but would limit the current in the diodes in line with LC traps along the coil. The diodes in line with the coil would then have large resulting equivalent series resistance (ESR), limiting their effectiveness. By placing four diodes in series at the LC balun output both in-line and LC balun diodes were sufficiently forward biased to have low ESRs in Tx. Matching and preamplifier decoupling: lattice baluns are used for matching the coil to 50 Ω. The capacitors $C_{tR}$ tune the coils to resonance. In addition, the lattice balun transforms the low impedance of the preamplifier, connected by $\lambda/2$ cables, to a high impedance providing preamplifier decoupling (42). The preamplifier (WanTCom WMA60A series) had a -3 dB gain bandwidth of 50-75 MHz and a -10 dB output reflection coefficient BW of 53-72 MHz. The preamplifiers also have a peak power limit of 30 dBm. Additionally, for both $^{1}$H and $^{19}$F frequencies the input reactance was tuned to be $\sim 1.5\ \Omega \pm 3i\ \Omega$ and the noise figure was measured to be $\sim 1.5$ dB with a 50 Ω load at both frequencies. Therefore, the preamplifier was deemed to be highly suitable for use at both frequencies.

The PCB with affixed rigid FR4 board prevents bending in the longitudinal direction and with foam padding ensures that there is at least 1.5 cm spacing between cables and the volunteers to prevent excessive tissue heating. Coupling between Rx coil cables and the coil itself during Tx was mitigated by running cables along the central axis. During operation, foam padding was affixed to protect the coil circuitry from damage.

**Tx/Rx Decoupling: MEMS vs PIN diodes**

Tx/Rx decoupling is essential for protecting the Rx circuity (the preamplifiers have a maximum peak power of 30 dBm) and preventing unwanted currents on Rx cabling. Here, two methods of Tx/Rx decoupling were tested, as shown in Figure 3a: one utilizing MEMS (MM7100, MenloMicro, Irvine, CA, USA) and one utilizing PIN diodes. For the MEMS Tx/Rx detuning, the switched open capacitance (~2 pF) provides a high blocking impedance, while the Tx/Rx decoupling using PIN diodes with LC traps is a commonly used and well-characterized method (43). The use and performance of MEMS is still relatively sparsely covered in the MRI literature, therefore, the decoupling provided by MEMS was compared to the PIN diode method, and the operation in the scanner was tested in practice at high power by MR imaging experiments.

The operation of the MEMS used are fundamentally different from PIN diodes as physical contact between conducting elements is made by the electrostatic force caused by a voltage applied between a gate and the conducting elements (shown in Figure 3a), the switch is actuated when the beams make contact with the central conductor. Recent improvements in the technology involving the use of a large array of MEMs with improved driver circuitry has allowed improved switching speed and reduced arcing, effectively eliminating these downsides seen in early applications of MEMs. More details on the device structure are provided in Figure 1 of reference (44). A comparison of the equivalent series resistance of the MEMS in switching has also been made in reference (19).

**MEMS Switching Voltage Up-conversion**

The particular MEMS switch employed requires 82 V between the beam and gate to transition between the "on" and "off" state. However, the GE HDx MRI scanner only provides a -3 V (in Rx) → +5 V (in Tx) DC bias voltage on the signal line of the Rx channels to control Tx/Rx switching. Furthermore, the high voltage (82 V) must be provided during Rx and the low voltage (0 V) provided during Tx. To drive the MEMS method of switching, the MRI system electronics, therefore, must be modified and high voltage switching functionality must be added.

The simplified circuit schematic shown in Figure 3b shows the block diagram/schematic of the method devised here to convert the control voltage from -3 V → 5 V to 82 V → 0 V. An inverting driver (MCP416T) converts the -3 V → 5 V control voltage to a 10 V → 0 V control logic required by the FAN7085 integrated circuit. The constant 10 V source provided by the system is required to operate the MCP416T and FAN7085 and is up-converted to 82 V utilizing bootstrapping provided by the FAN7085 high-side gate driver. The FAN7085 integrated circuit then converts the 0 V → 10 V output to a 72 V → 82 V, which drives a BSR92P BJT to regulate the final voltage output to the MEMS switches of 82 V → 72 V.

**MEMS Performance Testing**

The operational consistency of MEMS switching was tested using a test circuit consisting of a bias signal that emulates the scanner system default signals, the MEMS driver and a RF source. The bias signal and RF source (2 MHz voltage signal delivered to the RF pins of MEMS) were provided by a WS8352-Taber waveform generator, while a Keysight DSO 104a oscilloscope was used to measure the output. MEMS switching was also measured while scanning on the 1.5T GE HDx signa system to verify the operation. Testing in the MR system consisted of using the oscilloscope to measure the induced RF and DC voltages on the Rx input of a single coil (across the ports of the $D_s$ diode in Figure 3a, but with the $D_s$ diode removed), the Tx input to the 8-element array and on the MEMS DC bias line at the point of connection on the coil array.

**EM Simulation**

The finite-element method solver software HFSS was used for full-wave electromagnetic simulation of the impedance parameters, right circularly polarized Tx field ${B_1}^+$ and the left circularly polarized field per-unit-current $\widehat{B_1}^-$ (using current sources in place of lumped ports) for each Rx element. Coils are tuned, decoupled and matched using EM simulation. The Tx field, ${B_1}^+$, is simulated with the tuning capacitors of the 6-element Rx array removed so that only the 8-element Tx/Rx array is operational. Thus, the simulation of Tx efficiency was performed with ideal power division/excitation. Simulation of Tx efficiency was performed at 64 MHz and 60 MHz in a cylindrical phantom with dimensions and dielectric properties ($\varepsilon_r =$ 76 and $\sigma =$ 0.8S/m (45, 46)) matching the measurement phantom. The reported ${B_1}^+$ was normalized for a 1 kW RMS input power. The simulated $\widehat{B_1}^-$ was used to calculate the intrinsic SNR (iSNR) with the coil either acting as a single Tx/Rx coil or as Rx array with the iSNR calculated using:

$$iSNR = \frac{\boldsymbol{w}\hat{B}_1^-}{\sqrt{\boldsymbol{w}\boldsymbol{\Psi}^{-1}\boldsymbol{w}^{\mathcal{H}}}} \tag{1}$$

where $\boldsymbol{w}$ are the element weighting factors and $\boldsymbol{\Psi}$ is the covariance matrix (47). To emulate a single coil the fields were combined with a fixed phase corresponding to the opposite circular polarization of that in transmission, while for a Rx array the weights for optimum combination used are $\boldsymbol{w} = \boldsymbol{\Psi}^{-1}\widehat{\boldsymbol{B}}_1^-$(48)

The open-circuit covariance matrix is calculated from the simulated impedance matrix as $\boldsymbol{\Psi} = \mathbb{R}(\mathbf{Z})$, where $\mathbf{Z}$ is the impedance matrix (49). The optimally combined iSNR of the array was compared to the ultimate SNR (uSNR) as calculated in references (28) within a phantom equal in size to that of the simulated cylindrical phantom. The uSNR was calculated at 64 MHz to match the measurements at the $^{1}$H frequency, but results should apply reasonably well due to $^{19}$F imaging due to the close proximity of their respective Larmor frequencies. The fields from a plane-wave basis set, with the first 400 plane-wave field modes, were used to calculate the uSNR.

Noise introduced in the Rx chain by lossy passive elements, or added from active elements such as the preamplifiers, can be approximated by increasing the diagonal elements of the covariance matrix, $\boldsymbol{\Psi}$**,** in equation (1) by an amount proportional to the noise figure as in reference (30, 31). A reasonable estimate of attenutation (5 dB/100 m at 60 MHz) is used to derive the noise figure of 0.08 dB for the $\lambda/2$ =1.62 m cables used here. The T-matching network used for the 8-element Tx/Rx array consists of two series capacitors and a shunt inductor (see Supporting Information Figure S1). The noise figure of this matching configuration is dominated by the loss in the inductors (Q factor ~ 130 at 60 MHz), since the capacitors have Q factors >1000. Therefore, for the $\lambda/4$ T-matching network, with a reasonable estimate of the loaded coil resistance of $\sim 16\ \Omega$, the noise figure found is 0.14 dB. The $16\ \Omega$ coil resistance is an intermediate value between the value of the 8-element Tx/Rx array and 6-element Rx-only array. Details on the calculation of the noise figure for matching components that are in "shunt stage" configuration, such as the inductors here, are provided in reference (50). Including the noise figure of the preamplifiers (1.5 dB) the combined noise figure of 1.72 dB is estimated for all the receive elements in simulation. Thus, the overall noise figure is dominated by the preamplifier, and small differences in the noise figure of the matching network or cables would only have a small impact on the comparison performed here.

**Phantom Imaging**

Coil Rx and Tx performance was evaluated with the designed coil operating as a 14-element Rx array, 8-element Tx/Rx array, or as a single Tx/Rx element (fixed phase combination). In fixed phase combination routing for a single channel Tx/Rx coil was applied to the 8-element Tx/Rx array elements with the Rx cables and preamplifiers removed and replaced with a single preamplifier and 90° quadrature signal combining hybrid. A home-built cylindrical phantom (15.5 cm radius and 42 cm height) was used consisting of 3.6 g/L NaCl and 1.96 g/L $CuSO_4{\cdot}5H_2O$ salt solution (46) to represent a human thorax load. Due to the matching, tuning and preamplifier decoupling strategies employed, the coils matching and tuning did not have to be altered for the different test cases. For the purpose of this study the Rx sensitivity and Tx efficiency in the phantom were compared at the $^{1}H$ frequency (64 MHz), which should be closely applicable to the $^{19}F$ frequency (60 MHz) due to the relative proximity of the two Larmor frequencies. Mismatch at the $^{1}H$ frequency was expected to slightly reduce the Tx efficiency and homogeneity (19). For example, for a -6 dB reflection coefficient at the $^{1}H$ frequency (which is close to the measured value), a 12% loss in efficiency may be expected. However, the level of difference in the preamplifier noise figure due to mismatch was marginal (51). Flip angle (FA) maps from the cylindrical phantom were generated by imaging using a 3D spoiled gradient echo sequence (SPGR) (Table 1) by varying input power (8 equal steps in FA, up to the maximum achieved FA of 42° with a peak input power of 2kW, as shown in Table 1). Then, the signal was fit to the SPGR signal equation as in Reference (52). The $T_1$ was previously measured as 39.5 ms. The standard deviation of noise in the images was measured in a signal-free region of interest greater than 100 pixels. The image SNR (assuming signal from a fully recovered longitudinal magnetization) was found from fitting, and was used to represent Rx sensitivity.

The FA in the cylindrical phantom was related to the Tx efficiency by measuring the output voltage duration, shape and amplitude, while adjusting for cable loss (1.9 dB); similar to as described in reference (53).

Optimal SNR reconstruction of images was performed using the individual coils pixel intensities from each receiver and the noise covariance matrices calculated from measurement made with no RF excitation, as in reference (45). Noise covariance matrices were calculated from measurement of the signal received in all elements with no RF excitation.

### In-Vivo Imaging

In vivo lung imaging (see acquisition parameters described in Table 1) with inhaled $C_3F_8$ gas mixed with 21% $O_2$ (BOC, UK) was performed in two healthy adult volunteers (29 and 25 years old) with the 14-element array following informed consent and a protocol approved by UK National research ethics committee. Prior to breath-hold, 4 deep inhalations were taken of the mixture to fully saturate the lungs. At 1.5 T optimized SSFP imaging of $C_3F_8$ can provide improved SNR for in-vivo lung imaging over the commonly used SPGR sequences by a mean factor of ~1.7 (9), and was therefore employed here. Additionally, for a more accurate comparison, a Hamming filter was applied to k-space as in references (54), so that SNR was higher and could be accurately measured throughout the lung region. FA mapping was performed in-vivo by acquiring three successive 3D SPGR sequences within the same breath-hold and nearly identical imaging parameters, but with varying RF pulse amplitude (a peak input power of 2 kW for the 77.4° FA achieved) in Table 1. The actual FA was subsequently calculated pixelwise. Finally, using the Tx/Rx array co-localized $^{1}$H anatomical (15-second breath-hold) and $^{19}$F ventilation imaging (the average of two 16 second breath-holds) was achieved.

## Results

### Bench Coil Characteristics

The mean $Q_{unl}/Q_{load}$ of the anterior-posterior and left-right coils of the 8-element array was 230/20 and 158/15, respectively. The mean $Q_{unl}/Q_{load}$ of coils in the 6-element array was 283/26 In Rx, with the Tx/Rx switching PIN diodes unbiased (see Supporting Information Figure S1), elements of the tuned 8-element array had low reflection loss ($S_{11}$<-15 dB) and low coupling ($S_{21}$<-15 dB) between superior and inferior coils. Coupling between overlapped coils was also <-15 dB, while the maximum coupling observed was between diagonal elements, such as the superior-anterior and inferior-left coils, and was ~-12 dB. Coupling between all opposing coils (such as anterior and posterior coils) remained <-25 dB in measurements with the dielectric cylinder.

The reflection coefficient at the Tx ports of the four superior-inferior pairs (measured at the input to Wilkinson power dividers) was <-20 dB at the $^{19}$F frequency (60 MHz), but was -5dB$\pm$0.5 dB at the $^{1}$H frequency (64 MHz). The reflection coefficient at the input to the 90° hybrid combiner was <-20 dB at the $^{19}$F frequency and <-12 dB at the $^{1}$H frequency, indicating that much of the reflected power at the $^{1}$H frequency is largely deposited into the 50 Ω resistors present in the two 180° power dividers and 90° hybrid combiner.

### Tx/Rx switching

For the elements of the 6-element Rx only array, the coupling between adjacent coils was <-20 dB at both $^{19}$F and $^{1}$H frequencies, and the reflection coefficient was <-20 dB at the $^{19}$F frequency and approximately -6 dB at the $^{1}$H frequency. In Rx, the strongest coupling between elements of the 6-element Rx-only array and elements of the 8-element Tx/Rx array occurred between the closest anterior/posterior coil of the 8-element Tx/Rx array and the anterior/posterior coils of the 6-element Rx array, which was<-12 dB. The difference in $S_{12}$ as measured through a pair of pickup loops when switching the 6-element coils from Rx (resonant and matched) to Tx (detuned) was measured as a minimum of 35 dB (when loaded).

**MEMS Performance Testing**

The performance of the MEMS driver and MEMS state switching is shown in Supporting Information Figure S2. The MEMS showed a ~15 μs delay when switching into Rx (low impedance) state and ~12 μs delay when switching to Tx (high impedance) state. In both cases approximately 4 μs of this period is attributable to the MEMS switching itself and the remaining delay to the driver circuitry.

The isolation between the preamplifier inputs with MEMS or PIN diode Tx/Rx decoupling and the Tx input is shown in Supporting Information Figure S2. At the $^{19}$F frequency, the MEMs and the PIN diode/LC trap methods provided 22dB and 27dB additional isolation, respectively. At the $^{1}$H frequency the MEMs and the PIN diode/LC trap methods provided 24dB and 22dB isolation, respectively. There is an additional 6 dB of isolation expected to be provided by the additional two stages of power division before the superior-inferior Tx port. The total isolation (>40dB) exceeds the peak power limit required to protect the preamplifiers (<30dBm) with an expected peak input power of 2kW (63dBm). The measurement of the voltage induced on the preamplifier with a 2 kW Tx pulse was measured, and was significantly lower than that which would compromise the preamplifier operation.

**Transmit Efficiency and Homogeneity**

The simulated and measured Tx efficiency/homogeneity at the $^{1}$H frequency, in central coronal, axial and sagittal slices, are shown for the cylindrical phantom in Figure 4a and Figure 4b, respectively. The Tx efficiency maps are normalized to the mean Tx efficiency, measured as $B_1^+$ (μT), for a 1 kW RMS input power, found within the exemplified lung volume and the standard deviation of the Tx efficiency within the volume is quoted as well. The ~0.5 dB insertion loss for each of the three power divider stages, result in the lower Tx efficiency in measurement (6.79 $\mu T/\sqrt{kW}$±26%), when compared to simulation (7.45$\mu T/\sqrt{kW}$±20%). The lower homogeneity indicates minor coupling/loss with the Rx circuitry (cables, preamplifiers and MEMS detuning) and possible effects from changes in positioning between imaging sessions could also contribute.

**Receive Sensitivity**

Coil sensitivity maps are provided in Supporting Information Figure S3 for the 14-Rx channels. No significant coupling is evident between the coils as represented by the varied and largely non-overlapping regions of sensitivity, which is further corroborated by the low noise correlations between individual channels (generally <25%, a maximum of 35.6% and mean of all interactions of 10.6% as measured with the FA mapping imaging dataset).

The measured Rx performance for central axial and coronal slices of the cylindrical phantom are shown for the fixed phase combination, 6-element Rx array, 8-element Tx/Rx array configuration and for the total 14-Rx elements in Figure 5a-d respectively. The simulated Rx sensitivity, characterized by equation (1), for the same four configurations, are shown in Supporting Information Figure S4.

The measured mean increase in SNR using the 8-element array over the fixed phase configuration within the outlined lung volume was a factor of 3.3. Also, in simulation, the use of all 14-element resulted in a further 1.41 factor increase that closely matches the measured increase of 1.49.

A significant increase with the 14-element Rx array when compared to the fixed phase configuration is observed in the central region of the lungs, both in the measured results shown in Figure 6a and the simulated results shown in 6b. The comparison of the optimally combined SNR (considering noise added by the RF Rx chain) to the uSNR shown in Figure 6c, demonstrates that in central portions of the lung volume, ~70% of the uSNR is achieved with the 14-element Rx array. The projection of the SNR increase through the central coronal slice (labelled Ax1 in Figure 6a) is shown in Figure 6d, demonstrating clearly the improvement achieved at the periphery when compared to the center and specifically in the lung region. At the periphery significant increases in SNR may still be achieved with increased Rx element counts.

**In-vivo Imaging**

Figure 7a shows the results of in-vivo FA mapping with $^{19}F$ $C_3F_8$ ventilation imaging. The observed in-vivo FA homogeneity is 17.5%. $^{1}H$ structural and $^{19}F$ ventilation images are shown with coronal ventilation slices in Figure 7b, and coronal anatomical slices in Figure 7c. Central coronal, sagittal and axial slices for the other volunteer are shown in Figure 7d and demonstrate the SNR profile obtained throughout the body.

## Discussion

### Bench Coil Array Characterization

Complete elimination of all sources of coupling in practice is difficult due to the close proximity of elements required for good homogeneity. In this study, the maximum level of coupling between elements in the transceiver array was -12dB, which can be compared to values reported in recent publications of as high as -7.8 dB (55), with typical values of coupling lying in the range of -10 dB to -16dB (56-61). Thus, the level of coupling in this array design was found to be a reasonable level when compared to equivalent state-of-the-art designs. In addition, in Rx arrays it has been shown theoretically (62) and experimentally (63),(41) that levels of moderate coupling have a negligible effect on image SNR.

### MEMS Performance and Tx/Rx Decoupling

The 4 μs delay of the MEMS switch used here is minimal, but the driver circuitry employed adds an additional 11 μs delay. This delay is small when compared to the 50 μs reported in earlier performance testing of MEMS for MR applications (22), and is sufficiently short for the ultra-short TE (UTE) and zero TE (ZTE) sequences that may be useful for imaging $^{19}$F gases (54, 64, 65) and $^{1}$H signal in the parenchyma (1, 66, 67) in the lungs. Consequently, the method of using MEMS with the MR system established here introduces many new opportunities for multi-tuned Tx/Rx coil design.

However, there are still several barriers to the routine use of MEMS in coil design. First, the MEMS employed here are still rare and expensive when compared to PIN diodes (>£200 compared to <£10 for the devices used here). Furthermore, the activation voltage is still much larger than that easily provided by most MR systems, which we overcame with a custom designed driver circuit that operated as a step-up voltage transformer. The MEMS employed here also have higher ESR and larger board footprint than the PIN diodes, requiring additional resistors and decoupling capacitors. Also, the reliability of MEMS is a well-known issue, since there are many potential methods of failure, stiction, electrostatic discharge related breaking and failure from “hot-switching” (68). Since the reliability of MRI coils for clinical and research purposes is often the most critical requirement, the use of PIN diodes may be preferred.

**Transmit Efficiency and Homogeneity**

Although Tx efficiency at the $^{1}H$ frequency may be reduced due to impedance mismatch, the overlaid images of $^{19}F$ and $^{1}H$ confirm that the coil array can successfully be used for obtaining co-registered images of both nuclei in the same exam without adjustment. The ability to image both $^{1}H$ and $^{19}F$ provides the SNR benefits of a Rx array at both nuclei, while avoiding the cumbersome alternative of multiple Tx and Rx coils for both nuclei. A cylindrical phantom was used in simulation as a reliable geometry with known electrical properties that could be accurately compared to measurement with the constructed coil. However, in-vivo FA mapping shows that the expected Tx homogeneity and efficiency will be better than that observed with the cylindrical phantom due to the lower dielectric distortion in the human body because of the lower permittivity and geometry that is more conformal to the array geometry. Regions of low FA and signal to the inferior of the lungs are most likely due to the volunteer raising their arms above their head during imaging, thus positioning the array higher on the torso than in simulation.

**Receive Sensitivity**

The Rx sensitivity and signal reception profile homogeneity of the combined 14-element Rx array was significantly improved when compared to the single element equivalent configuration of the 8-element array. Thus, the combination of a Tx/Rx array with Rx-only array was found to be beneficial even when imaging at the relatively low frequencies used here, as they can address the need for coils to be close to the subject, while allowing for geometrical flexibility and avoiding the use of tight-fitting fixed-geometry coil housing.

The measured mean SNR increased by a factor of 5.1 with the designed 14-element array when compared to a fixed phase single transceiver configuration, which will greatly improve the quality of $^{19}F$ lung imaging routinely obtainable. However, the increase at the inner region of the lungs is lower; about a factor of 3.7 for the lowest increase shown in Fig 5c. Also, the noise figure of the preamplifier, with contribution from matching network and cables, limits the SNR improvement by at least the value of their combined noise figure (estimated at 1.72 dB). Previously, it has been shown that the simulated Rx sensitivity of array may remain the same with increasing element count when noise contributions from the coils or other sources are included (29). Thus, if the noise figure of the components such as the preamplifier could be reduced, the 14-element array shown here would likely be comparable to the Rx sensitivity of even larger element count arrays.

In phantom simulation, in a region equivalent to the center of the lungs, the 14-element array was found to reach ~50% of the uSNR (similar to observed with 8-16 element arrays in reference (69)). In previous works, a potential 80% of the uSNR was achieved by increasing the number of elements to >32 (30). Therefore, for applications that may require looking at the periphery of the lungs increased element count may still be beneficial. However, the imaging resolution for diagnostic images is limited by the lowest SNR region obtained, since ventilation defects must be detected with the same precision throughout the lungs. For the application of $^{19}F/^{1}H$ imaging at 1.5T, further increases in array element count are expected to provide little benefit and result in added engineering complexity and cost with an increased risk of component failure. If additional Rx elements were desired, elements orthogonal to the loop coils (45, 70, 71) could be added with no detriment to the Tx or Rx efficiency of the current design, though space may be limited and this would add to the complexity of cable management. Alternatively, a new transceiver array and/or Rx array could be designed with more elements that are reduced in size in order to cover the same field of view, though this would also greatly add to the complexity of the design.

## Conclusion

An 8-element flexible transceiver array with an additional 6-element Rx-only array was presented for imaging of $^{1}H$ and $^{19}F$ in the lungs at 1.5T. Through simulation and experiment this array was shown to provide significantly higher SNR than that of a single channel coil, and the addition of a 6-element Rx array also improved SNR in the center. Furthermore, for the first time, this study demonstrates the use of MEMS in the Tx/Rx detuning of a Tx/Rx array, which allowed the array to be used for $^{19}F$ and $^{1}H$ MRI. The only trade-off found for the single tuned-array operating at both frequencies was reduced Tx efficiency at the $^{1}H$ frequency. The improvement in Rx sensitivity measured experimentally was similar to that simulated. Additionally, simulation results showed that in the central part of the lungs (bordering the heart) >70% of the uSNR is expected to be achieved. Therefore, we believe the results presented here are close to the maximal $C_3F_8/^{19}F$ ventilation image quality obtainable at 1.5T with optimized sequence and RF coil hardware design, as limited by the central lung region.

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## Table Captions

Table 1: Imaging parameters for phantom and in-vivo SSFP and SPGR performance verification with $C_3F_8$

## Figure Captions

Figure 1: a: A diagram of the conductor traces for the 6-element receive array and 8-element array with the dimensions labeled. The simplified circuit schematics of the **b:** 8-element Tx/Rx superior-inferior pairs and **c:** three elements of the 6-element array is also shown. The location of LC traps for Tx-Rx detuning and lattice balun for coil matching are visible for the 6-element array**.**

Figure 2: The simulation model (left) and 3D CAD model (right) of the wrapped PCB for the 8-element Tx/Rx array and 6-element receive-only array together are displayed. The Tx/Rx circuitry is enclosed in the white boxes.

Figure 3: In **a:** two methods of isolating Rx circuitry during Tx are compared and the circuit diagrams are shown. In the first, PIN diode forward biased producing a high impedance with the LC trap and MEMS with switches open (MEMS diagram and circuitry for switching shown on the right). In **b:** a simplified circuit schematic is shown of the voltage up-conversion circuity to convert the 5 V control voltage to 82V.

Figure 4: **a:** Simulated and **b:** measured transmit homogeneity at the $^1$H frequency in axial, coronal and sagittal slices within a cylindrical phantom. The mean and standard deviation of transmit efficiency within the volume of a typical lung (shown as line art) is displayed above the axial slices. Sagittal slices are taken through the center of each outlined lung region.

Figure 5: The measured receive sensitivity for four different coil configurations in a cylindrical phantom for axial sagittal and coronal slices are presented as follows: **a:** the fixed phase combination, **b:** as a 6-element Rx-only array **c:** operating as an 8-element Tx/Rx array, **c:** or with the full 14-receive elements. The mean and standard deviation of the presented maps, within the outlined lung volumes, are labeled below the axial slices.

Figure 6: The ratio of simulated increase in receive sensitivity is shown for **a:** the measured and **b:** simulated optimum combination of the 14-element Tx/Rx array compared to the fixed phase combination. Additionally, **c:** the uSNR is compared to the simulated combined 14-element array. The mean and standard deviation of the presented maps, within the outlined lung volumes, are labeled below the axial slices. In **c:** the projection of the three cases along the coronal slice (line labeled Ax1 in **a:**) is shown along with transparent overlay indicating the region of the lungs.

Figure 7: In-vivo ventilation and co-localized anatomical/structural data sets acquired in the same imaging session with the coil operating as a 14-element array are displayed as follows: **a:** $^{19}$F FA maps, **b:** SNR maps of the $^{19}$F $C_3F_8$ ventilation imaging using a SSFP sequence acquired in two separate breath-holds and **c:** $^1$H anatomical/structural images measured with a 2D SPGR sequence. Also shown, **d:** ventilation images overlaid on anatomical images in central coronal, axial and sagittal slices obtained in another volunteer with the same imaging sequences.

Supporting Information Figure S1: In **a:** a circuit schematic of the Tx/Rx matching (shown in orange), Tx power splitting (shown in red), superior-inferior capacitive decoupling (shown in purple) and Tx-Rx decoupling circuitry (shown in blue) used for superior-inferior coil pairs. The connections to the conductors of the coils are shown as arrows (green for superior and blue for inferior), which are also shown for the same points in the **b:** constructed coil circuitry and in the PCB layout of Figure 2a.

Supporting Information Figure S2: **a:** The switching performance test of MEMS devices and driver circuitry together demonstrated with oscilloscope measurements of voltage waveforms across ports of MEMS. **b:** Transmission between Rx port of superior anterior coil, with preamplifier bypassed, and input of hybrid coupler feeding Tx ports of anterior coils with no detuning (baseline), or with detuning provided by MEMS or PIN diode detuning methods.

Supporting Information Figure S3: **a:** Individual sensitivity maps for coils shown as axial slices through the centre of each coil with a cylindrical phantom.

Supporting Information Figure S4: The simulated receive sensitivity for four different coil configurations in a cylindrical phantom for axial, sagittal and coronal slices are presented as follows: **a:** the fixed phase combination, **b:** as a 6-element Rx-only array **c:** operating as an 8-element Tx/Rx array, **d:** or with the full 14-receive elements. The mean and standard deviation of the presented maps, within the outlined lung volumes, are labelled below the axial slices.