Regional Faraday shielding for improved dynamic hyperpolarized ^{13}C MRI

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Target Audience: MR scientists interested in designing optimized protocols for preclinical and clinical hyperpolarized ^{13}C MRI studies.

Purpose: Hyperpolarized (HP) MRI is limited by irreversible decay of the HP state, demanding an optimized approach to retain magnetization. Spatially selective excitation (by selective pulse design and/or local RF coils) is useful for restricting expenditure to the immediate imaging FOV. In practice however, spatial selectivity is imperfect and especially problematic for refocusing pulses needed to exploit the long T_2's of ^{13}C probes (1). Imperfect RF homogeneity (e.g. at coil ends) particularly increases the challenges. In reality, inadvertent losses in accessory areas are inevitable if refocused data is collected during delivery to, and recirculation about, the immediate region of interest (2). Capturing the early signal dynamics including the input function can be critical for accurate quantitative modeling. **In this work we investigated a new hardware-based approach using metallic Faraday shielding around key areas to minimize extraneous losses of HP magnetization in dynamic studies due to premature saturation during tracer delivery.**

Methods: *Initial testing-* Regional RF shielding was produced by wrapping standard aluminum (Al) foil cylindrically around enclosed areas. A similar idea was proposed previously for ^{1}H MRA using thermally polarized water delivered in a "Faraday catheter" (3). A dual-tuned ^{13}C/^{1}H quad volume mouse coil (*l*=8cm, *d*=5cm) was used in a GE 3T scanner. A pair of 1mL 8M [^{13}C]urea vials (*l*=4cm,*d*=8mm) were positioned adjacently in parallel longitudinally in the fringe field, extending 2cm beyond the inferior end-ring. The vials were imaged with and without shielding around one vial (Fig. 1). This was repeated for 1x,2x,&3x layered foil (t≈15 μm), or ~1-3 skin depths. We tested the effect on tuning, and saw a max frequency shift of +80kHz. The foil thus produced a small effect on tuning, but the loaded input match was ≤-9.4dB on both quad ^{13}C channels before and after. *Simulations-* A constant, plug flow model of tracer delivery through a mouse tail was simulated (Fig. 2). The tail (*l*=6cm) was infused with boxcar input of unit amplitude, lasting 20s. During transit to the mouse body at velocity *v*=10-20cm/s, control tail magnetization was repeatedly excited as a function of position along the tail in the fringe field of the coil, by 5° excitation and a pair of adiabatic hyperbolic secant spin echo pulses for each TR, as previously described (4) (duration= 10ms, spacing= 35ms, TR=250ms, adiabatic threshold= 1.2G). Another simulated tail was shielded from these pulses. B_1 was scaled to a maximum of 1.4G at the proximal tail end, with tail field scaled along *z* based on quasi-static EM simulation of the fringe field of the uniform birdcage mode. The tail was assumed to extend 2cm into the coil to meet the mouse body. Magnetization was updated after each pulse in Bloch simulations. Spoiling was assumed after each TR. Net accumulated $M_z(t)$ reaching the body was computed. T_1 (46s) relaxation was included. *HP phantom studies-* 40 mM HP [^{13}C]urea was obtained by dissolution DNP. Two catheter extension tubes (*l*=91cm,*i.d.*=0.67mm) were run in parallel through the end of the RF coil, extending 2cm inside, emptying into separate fixed reservoirs. 1mL HP urea was drawn into a pair of syringes and advanced through each tube at identical velocity (15cm/s) by hand over 20s. The experiment was repeated 3x, with and without shielding (3x) on the terminal 6cm stretch of one tube. Dynamic imaging was initiated with start of injection, by single shot EPI with α=5° and adiabatic DSE, at 5x5mm^2 with parameters matching the simulations. At 30s the catheter was flushed with .2mL H_2O to ensure delivery of same amounts of urea.

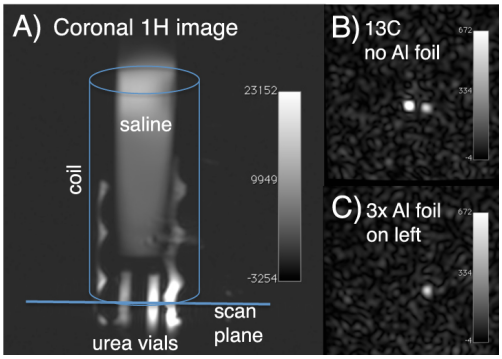

Fig. 1. Image of initial experimental setup for feasibility testing (A), and axial ^{13}C images with (C) and without (B) shielding on urea vial at left.

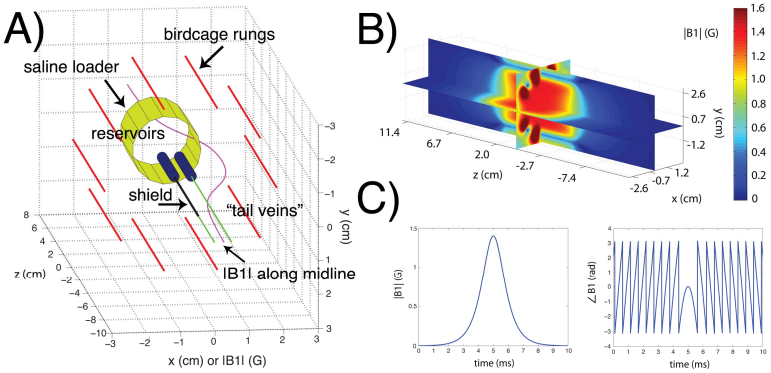

Fig. 2. Simulation setup showing shielded and unshielded lines extending into RF coil (A). B1 field distribution (B) and adiabatic spin echo pulse waveform (C).

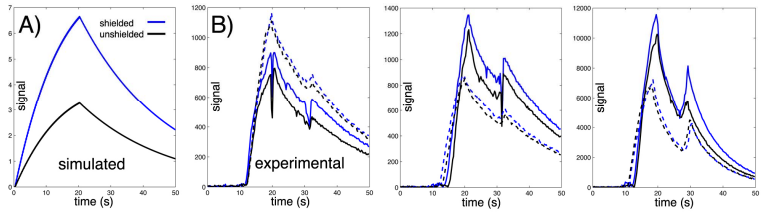

Fig. 3. Simulated (A) and experimental (B) signal for dynamic spin echo imaging during HP tracer delivery with (blue) and without shielding (black). Control experiments with no shielding on either tube shown by dotted lines.

Results: 1-3x foil all eliminated signal in the shielded urea vial, without affecting its partner (Fig. 1). Since images reflect a squared Tx/Rx effect, 3x foil was chosen to ensure full attenuation of transmission. No RF heating was seen. A large advantage was observed for tail shielding in simulations, especially at lower inflow velocities. At 15cm/s, the integrated shielded signal was increased 102% (Fig. 3A). Experiments demonstrated a more modest signal gain (Fig. 3B) of 21±2% (controls= 4±5%).

Discussion: Both simulated and experimental results demonstrate a clear advantage for shielded delivery of HP tracer for dynamic spin echo imaging, although experimental gains are significantly less than predicted. Considering that the shielding clearly works, errors may be due to factors related to infusion, such as flow pulsatility during infusion by hand, or deviation from plug flow. Other possibilities are that our fringe B_1 pattern is inaccurate or we have underestimated the B_1, potentially exceeding the adiabatic threshold in the fringe.

Conclusion: We have demonstrated feasibility of a new method for limiting losses of HP ^{13}C magnetization due to inadvertent excitation during delivery to a primary region of interest, using regional Faraday shielding.

References- 1. Reed et al. IEEE TMI. In press. 2. Josan et al. JMR. 2011. 3. Dumoulin et al. US pat #5419325. 1994. 4. Cunningham et al. JMR. 2008.