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BASICS OF NUCLEAR MAGNETIC RESONANCE I

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The "CIBM translational MR neuroimaging & spectroscopy" course

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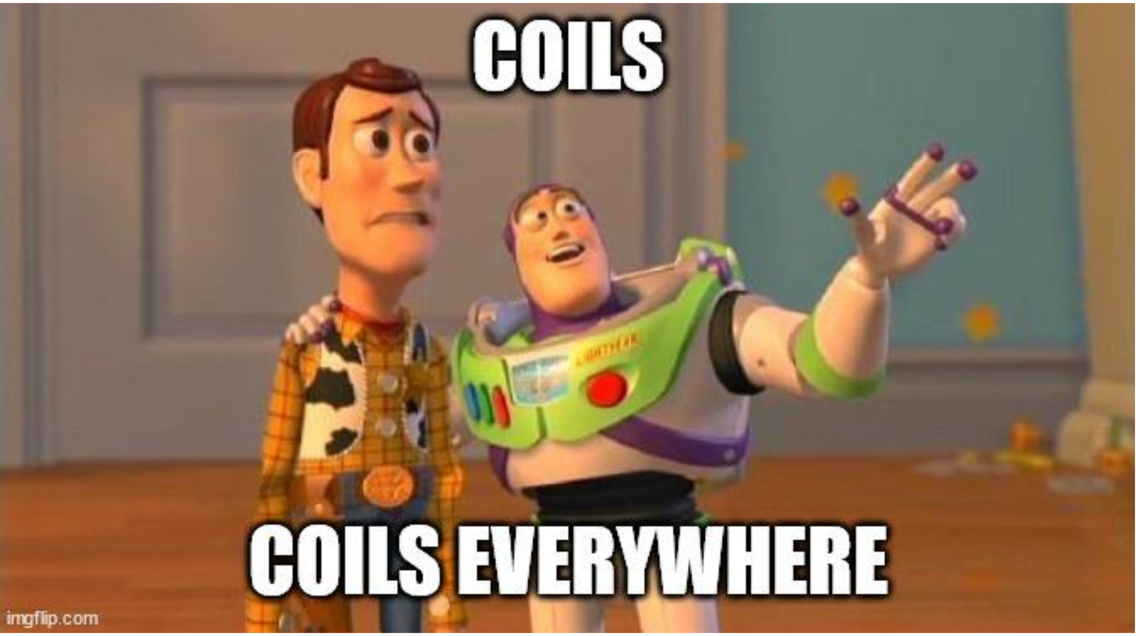
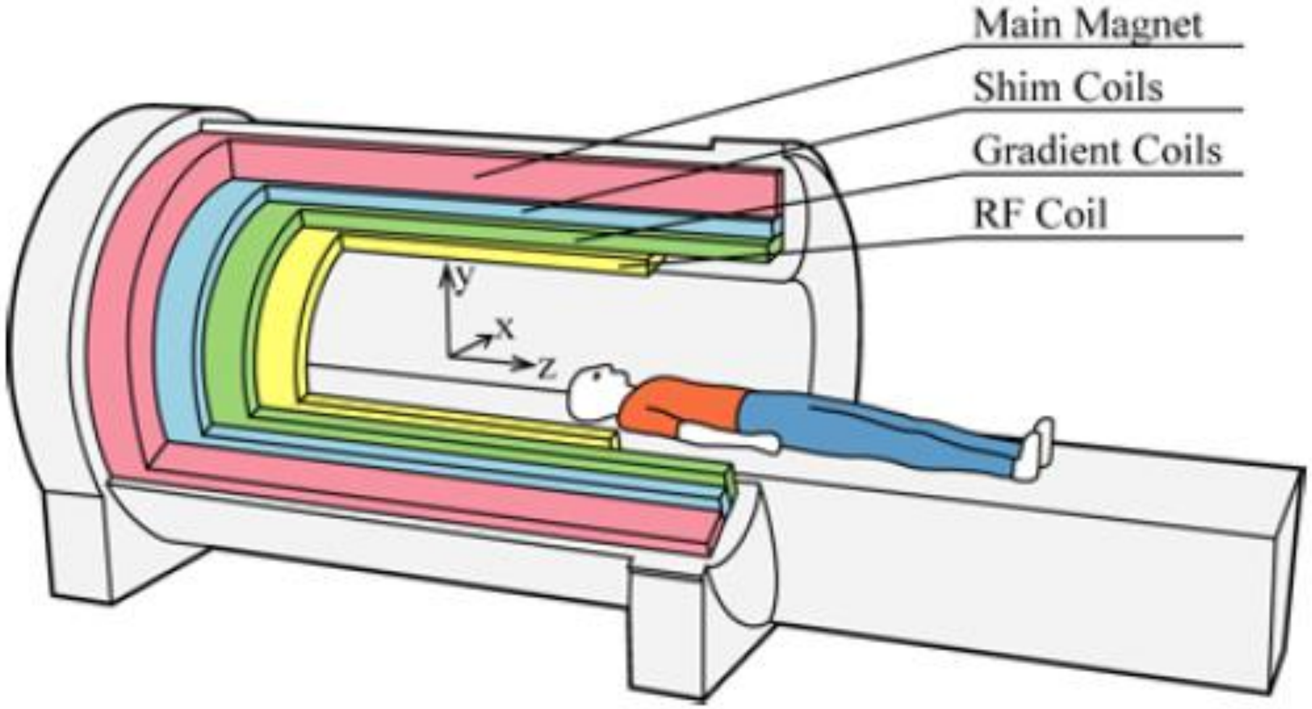
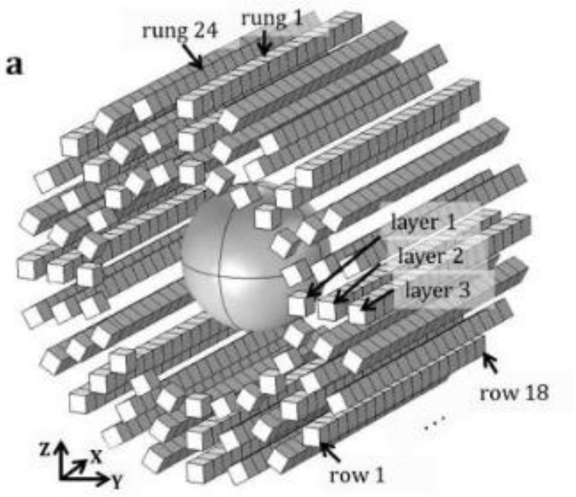
MR SYSTEM OVERVIEW





MAIN MAGNET

0.075 T

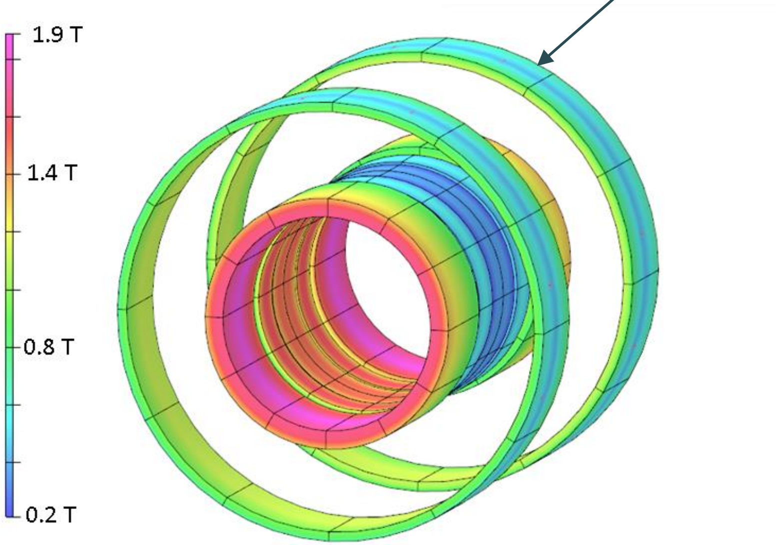


0.2 T

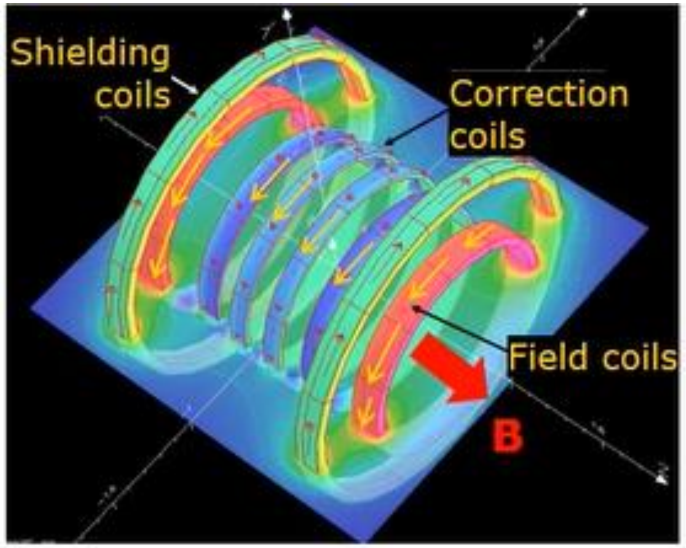
7.0 T

MAIN MAGNET – HIGH FIELD EXAMPLE

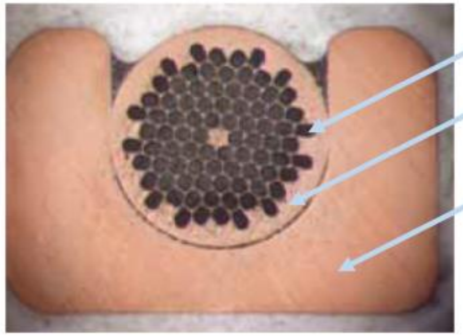
Active shielding against fringe field



Magnetic field on conductors to achieve 1.5 T in centre of magnet

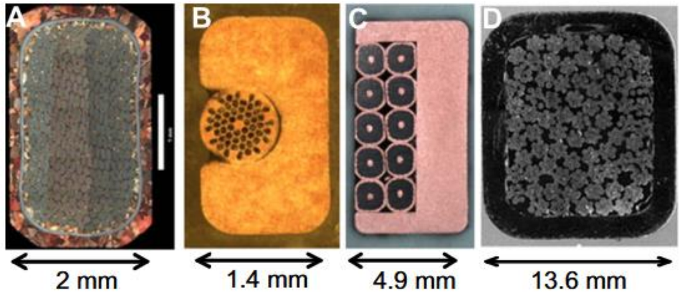


NbTi is typically used as a superconductor for MRI



Typically run at J_E = 100 – 150 Amm^{-2} current density

1.5 T magnet requires approx. 20 – 30 km conductor



B. Parkinson, ISMRM 2018; H. Fischer ISMRM 2018

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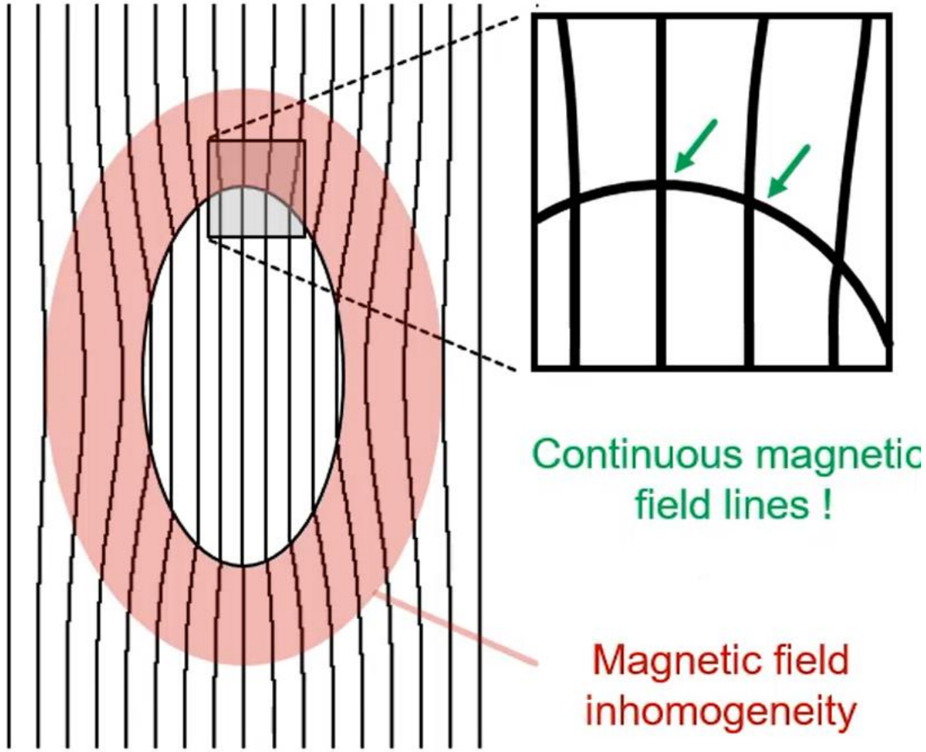
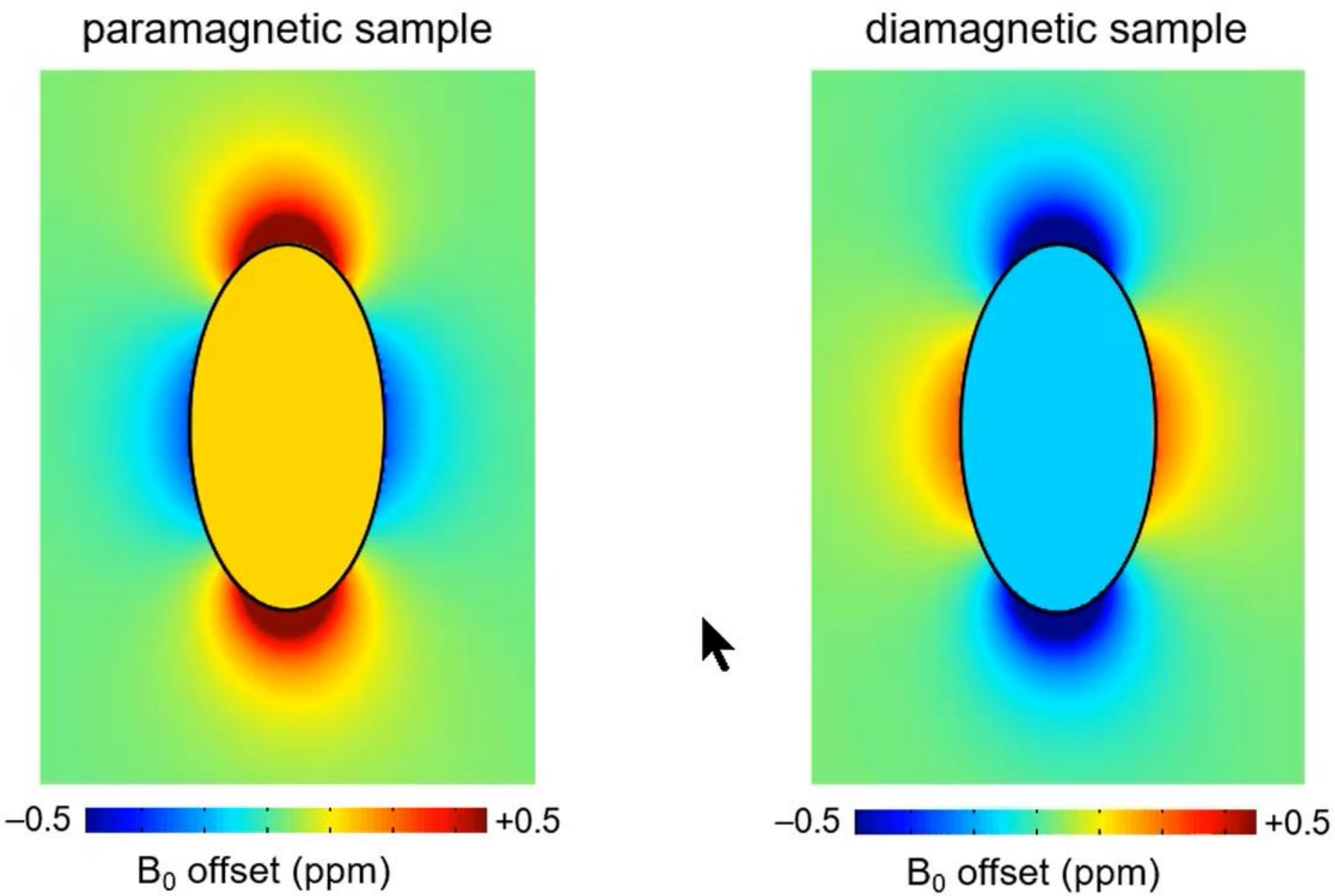
Fig. 5. Conductors used in superconducting magnets. A) Single-strand of Nb_3Sn wire from NHMFL 21.1 T / 10.5 cm NMR magnet; this wire carries 285 A at 21.1 T. B) NbTi Wire-In-Channel for a 1.5 T MRI magnet; this wire can carry ~200 A at 5 T. C) 10-strand NbTi cable with Cu stabilizer and reinforcement designed for 11.75 T / 90 cm magnet under construction by Iseult; this cable can carry 1,500 A @ 12 T. D) 525-strand Nb_3Sn cable with stainless steel reinforcement from NHMFL 45 T hybrid magnet; this cable can carry 10,000 A at 15 T. Photos in B and C courtesy of Hem Kanithi.

B0 INHOMOGENEITY

Paramagnetic sample



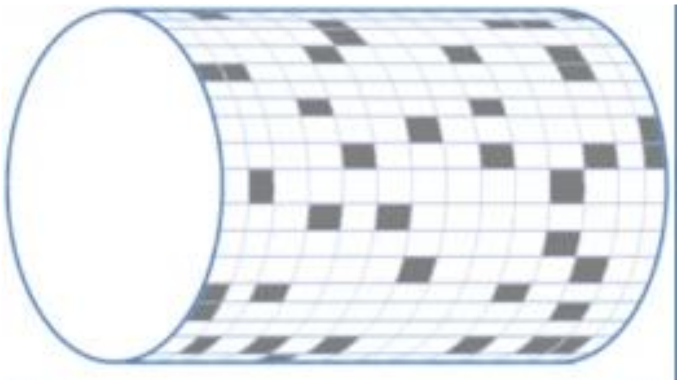


R. de Graaf, ISMRM 2019

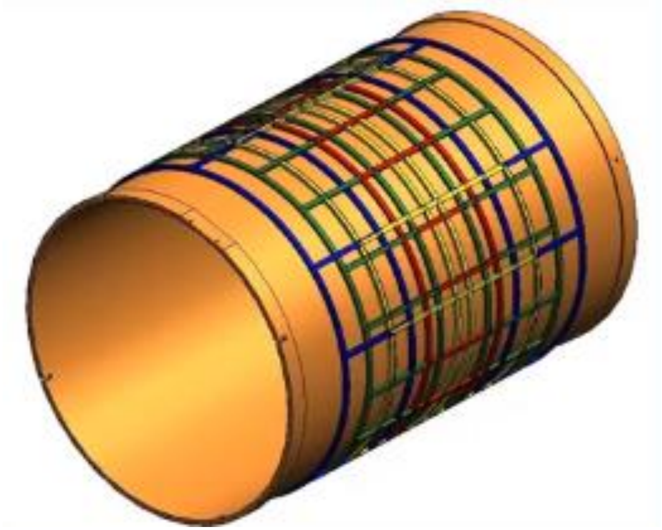
B_0 SHIM COILS

Diamagnetic/paramagnetic sample interacts with B_0 field

- Passive shimming
  - Ferromagnetic inserts

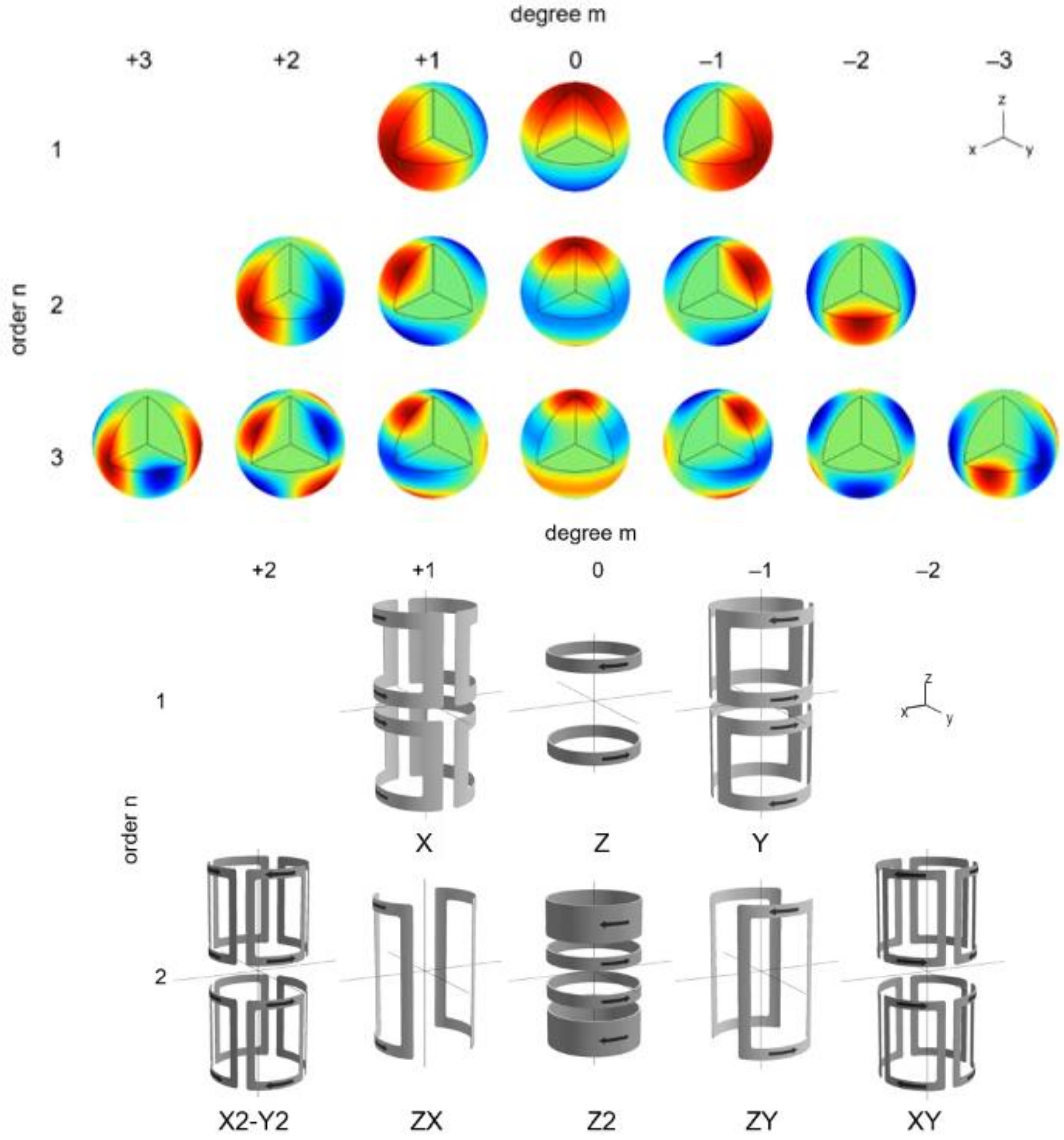


- Active shimming (superconductive, resistive)
  - Conductive loops carrying direct currents (spherical harmonics analysis)
  - $\Delta B_0(x,y,z) = B_{0,offset} + \sum_{n=1}^{\infty} \sum_{m=-n}^{+n} C_{n,m} F_{n,m}(x,y,z)$



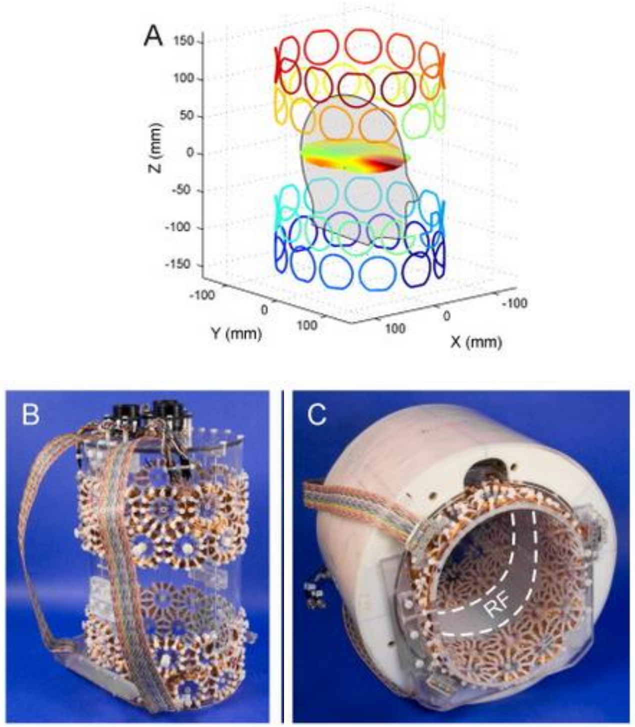
J. Overweg, ISMRM 2019; R. de Graaf, ISMRM 2019

Spherical harmonic functions



R. de Graaf, ISMRM 2019

Multi-coil dynamic shimming



Juchem et al. JMR 2011

Stockmann et al. MRM 2016

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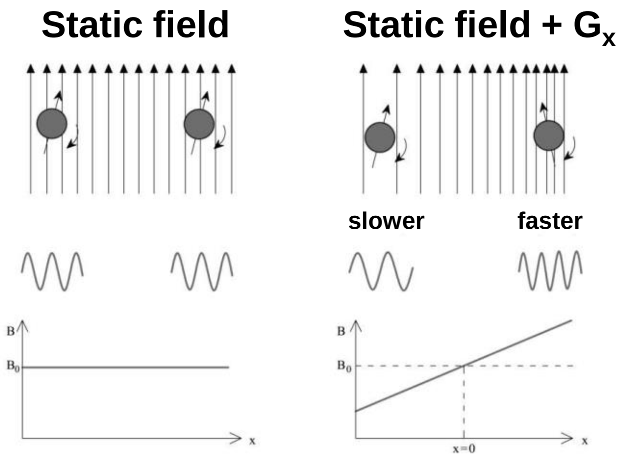
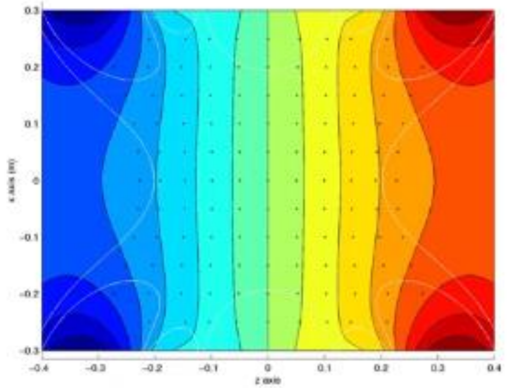
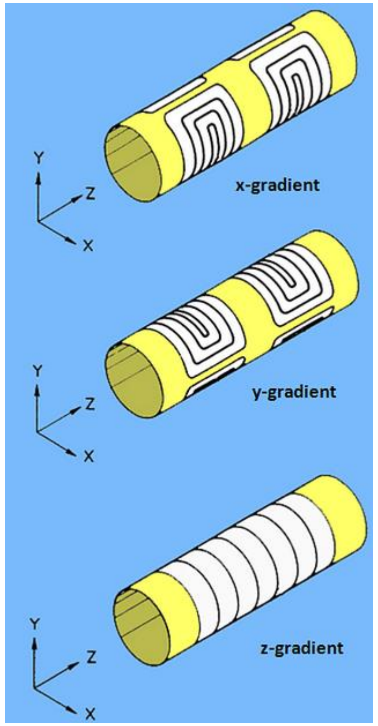
GRADIENT COILS

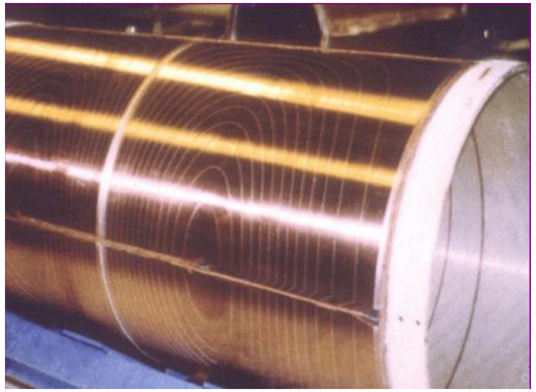
- Produce magnetic gradient fields in three directions

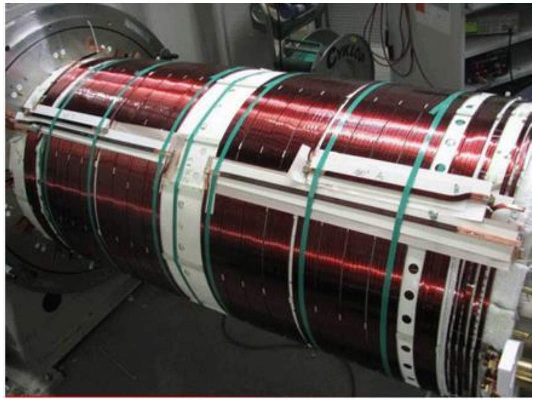




R. Bowtell ISMRM 2019







NUCLEAR MAGNETIC RESONANCE

- In MRI we need to produce a magnetic field B_1, that is:
  - perpendicular to B_0
  - oscillating at the Larmor frequency:

$$\omega_L = \gamma B_0$$

where γ is the gyromagnetic ratio.

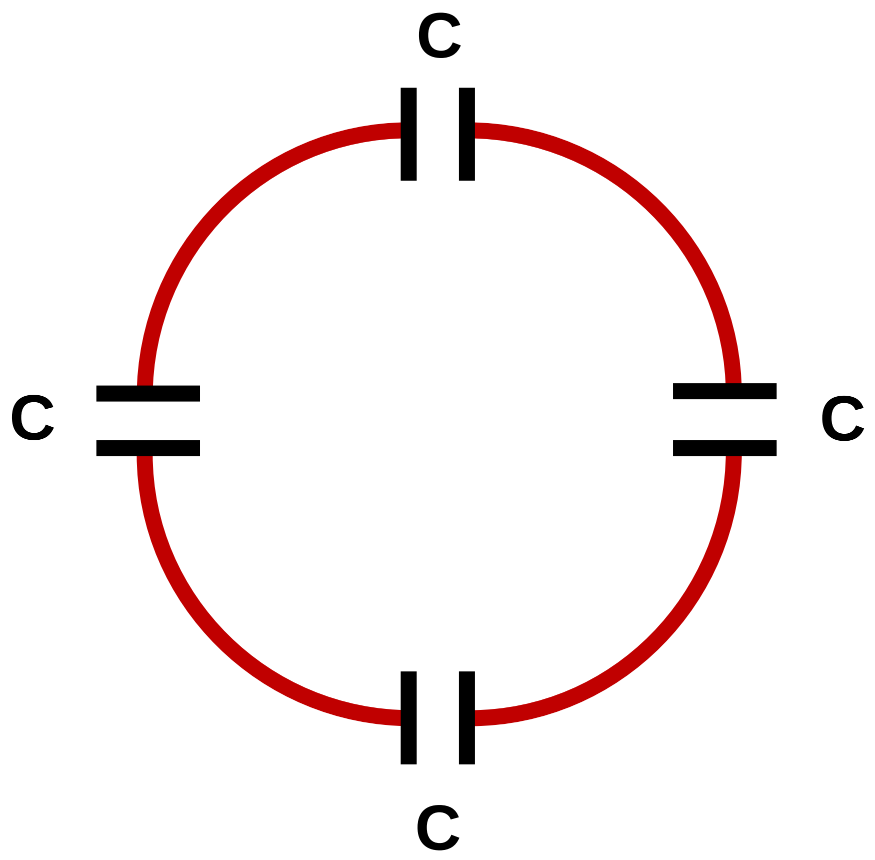
- For protons, γ = 42.6 MHz/T
- For field strengths 0.2 – 20.0 T, $f_L = \omega_L / 2\pi \rightarrow$ between 8.6 - 860 MHz

HOW TO PRODUCE THE B1 FIELD?

Radio frequency coil (RF coil)

INDUCTANCE, L

CAPACITANCE, C



$$\omega_0 = \frac{1}{\sqrt{LC}}$$

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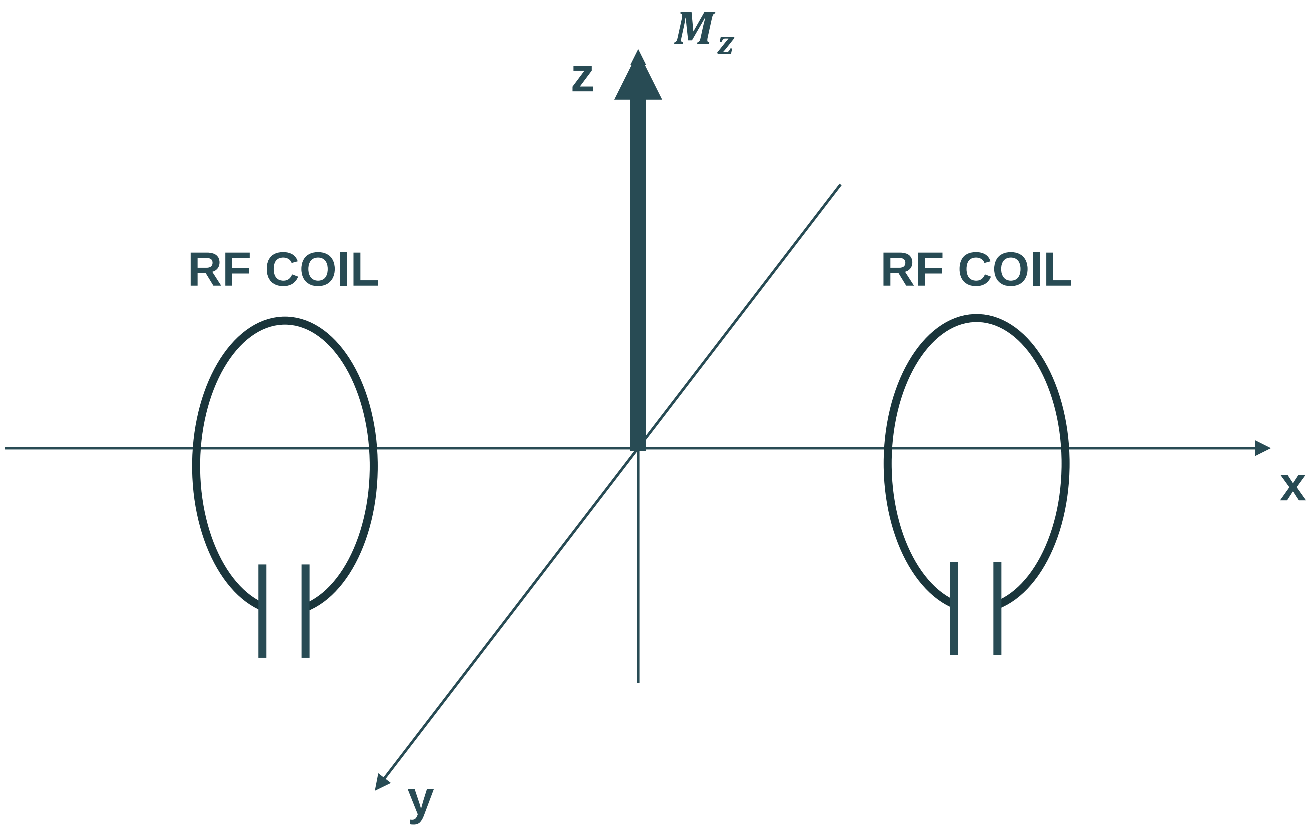
DUAL-FUNCTION OF RF COIL IN MRI

- **RF TRANSMISSION**
  - Tip of net magnetization
- **NMR SIGNAL RECEPTION**
  - Induction of voltage in the coil



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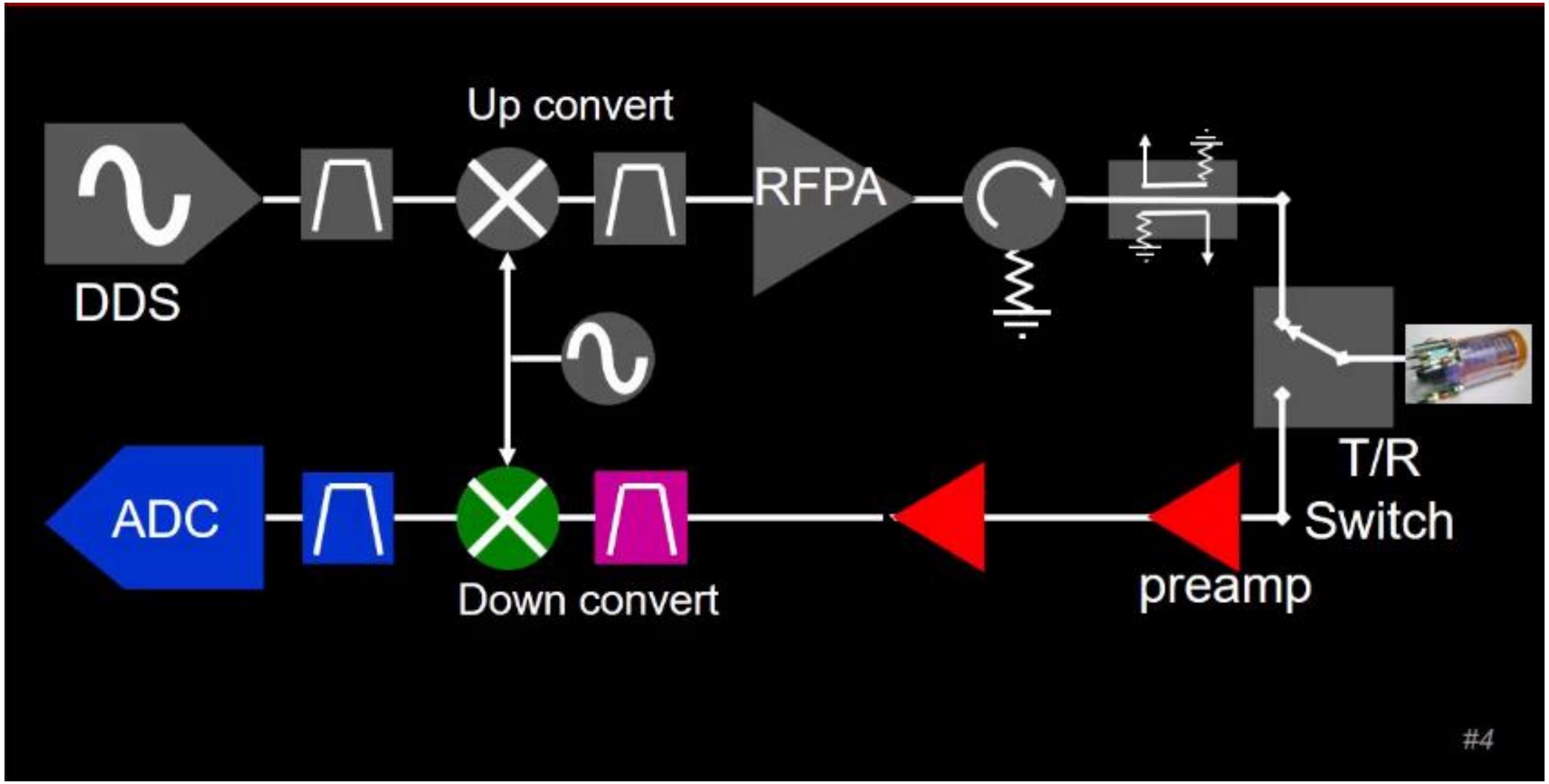
RF COILS

Different configurations possible:

- Transmit/receive (TXRX)
- Transmit-only (TX-only) + receive-only (RX-only)
- Transmit/receive (TXRX) + receive-only (RX-only)

- Can be combined with other RF coils tailored for other resonance frequencies (X-nuclei)

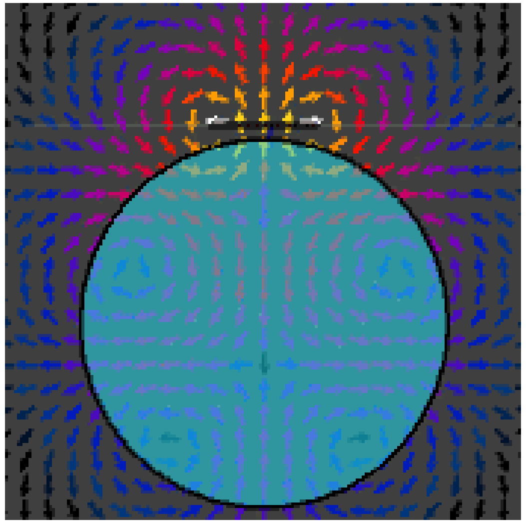
TRANSMIT/RECEIVE CHAIN

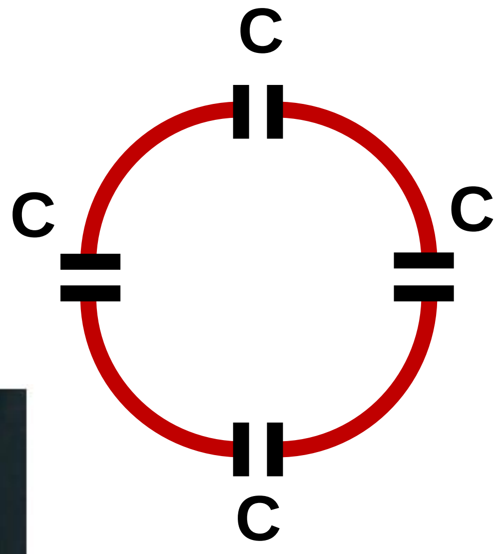


SURFACE COIL

- Easy to build
- High SNR
- Heterogeneous B_1 field
- Either TXRX or RX-only
- Can be extended into an array

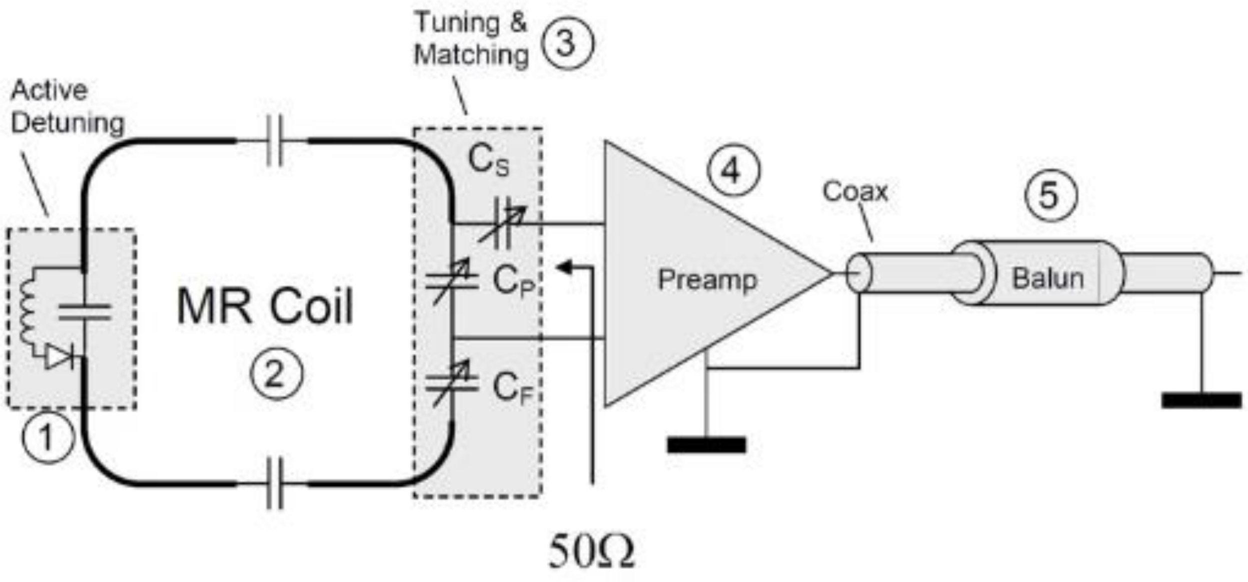
LOOP COIL
(H-field)





RECEIVE-ONLY COIL

Useful for all magnetic field strengths and applications – high SNR



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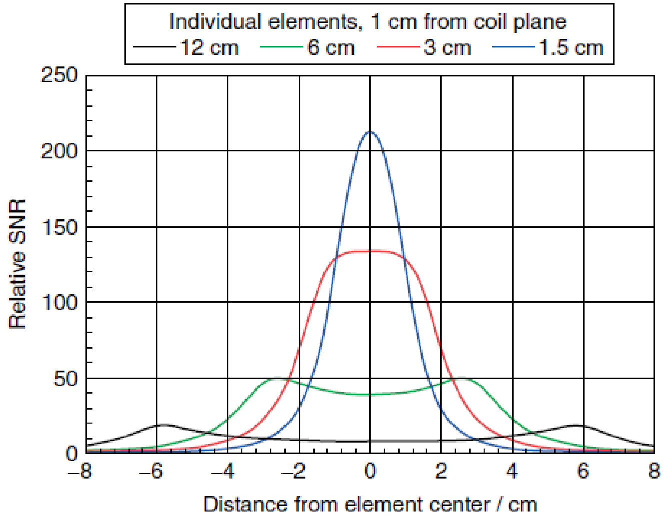
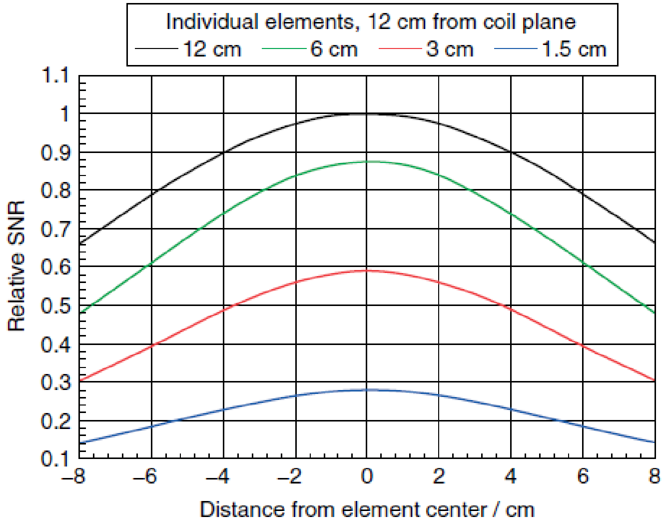
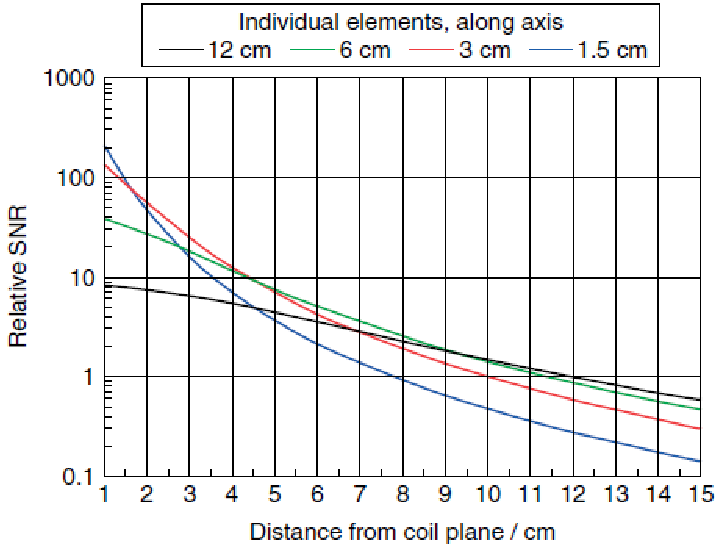
SURFACE COIL: WHICH SIZE?



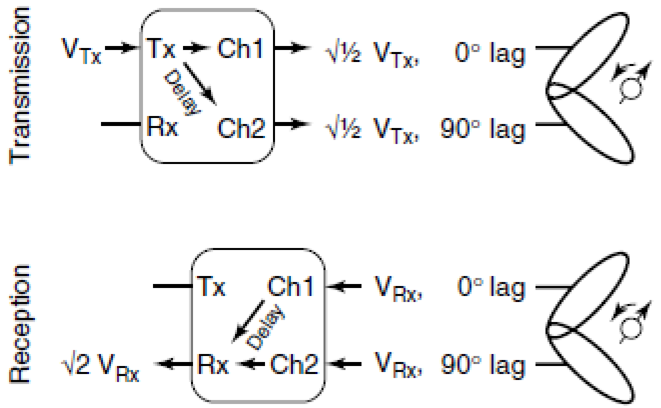




Wright, chapter in book "RF coils for MRI" (2012)

QUADRATURE

- Two waveforms with a 90° difference in phase are said to be in quadrature
- If these are used to drive two RF coils producing B_1 fields that are equal to each other in space, an RF magnetic field with circular polarization will result
- Gains:
  - Improves SNR by √2
  - Reduces transmit power by a factor of 2



Webb and Collins, chapter in book "RF coils for MRI" (2012)

QUADRATURE SURFACE COILS

- Higher transmit efficiency than single coil element
- Heterogeneous B_1 field
- Limited FOV
- Can be used in low and high field MRI

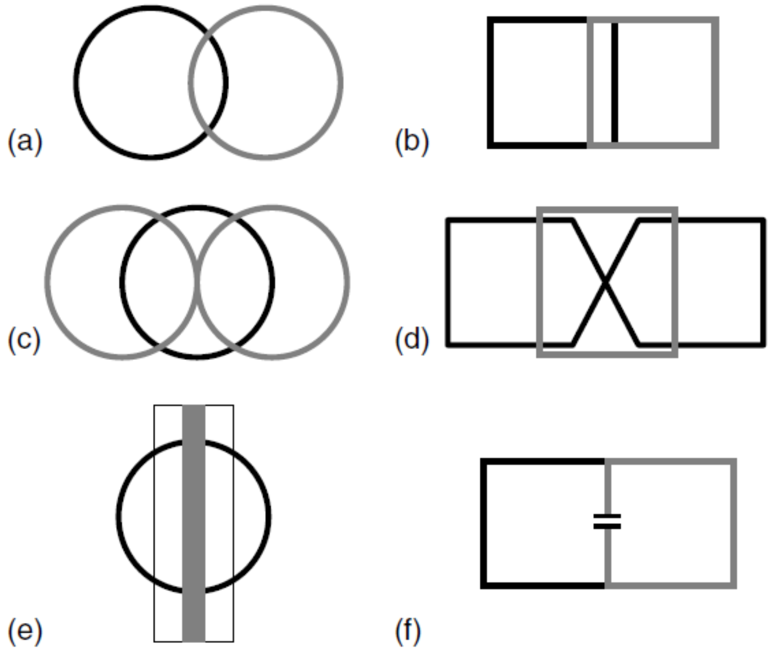
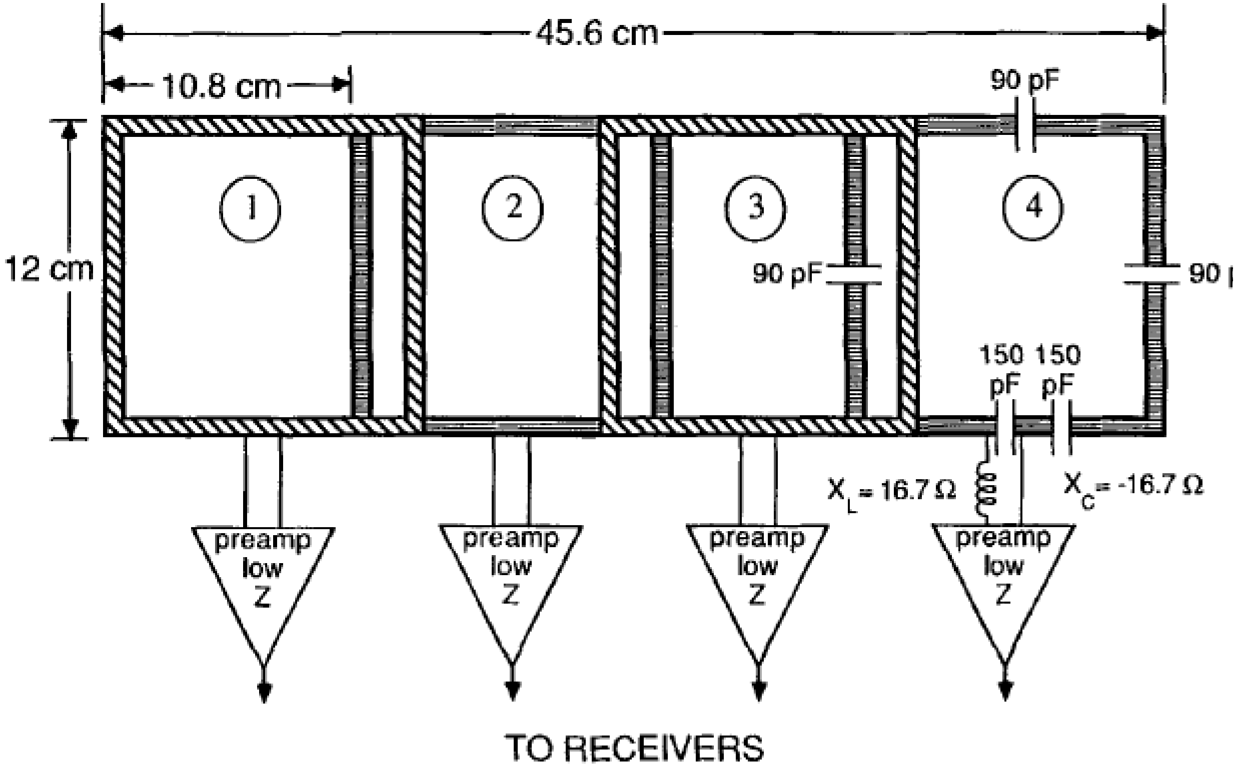


Figure 2 Different types of quadrature surface coils: (a) and (b) Two-loop structures with the overlap between the two coils chosen to minimize the mutual inductance. (c) and (d) Butterfly arrangements in which the intrinsic symmetry of the arrangement produces isolation between the coils. (e) A combination of single loop and stripline resonator. (f) Two loops with a single isolation capacitor between the two loops

MULTI-CHANNEL ARRAYS: SNR GAIN



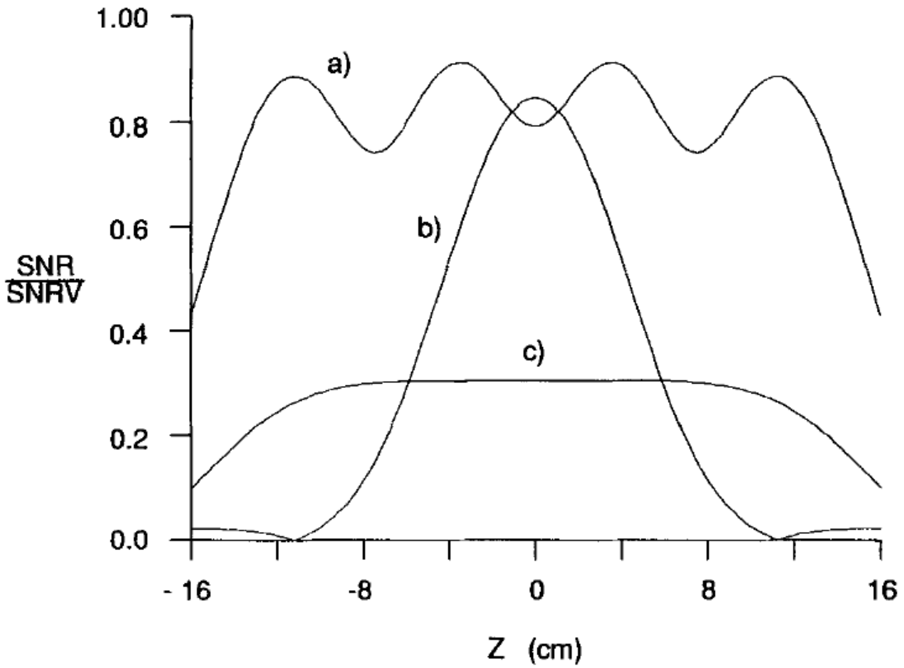
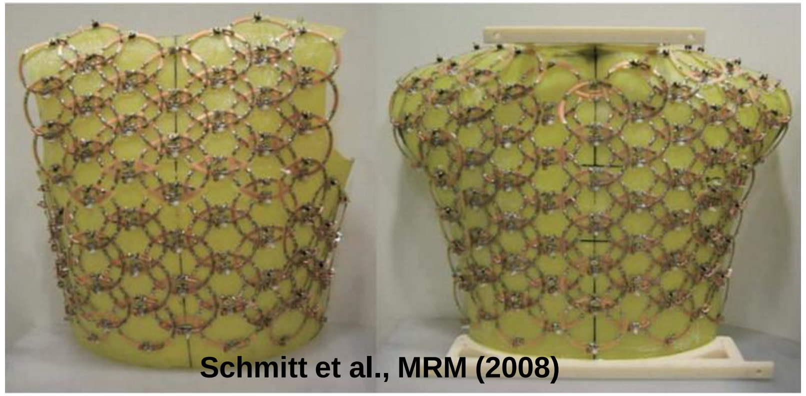


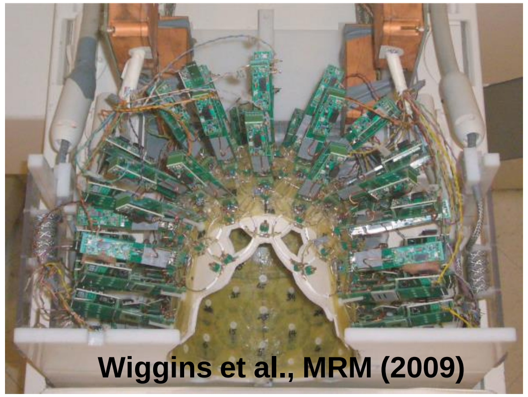
FIG. 10. Calculated SNR at a depth of 8 cm for three different surface coil arrangements: (a) a four element phased array made of 8-cm square coils; (b) a single 8-cm square coil; and (c) a single large 30 × 15-cm rectangular coil. An SNR of 1 corresponds to a theoretical upper limit SNRV given by Roemer and Edelstein (*4*) for linear reception.

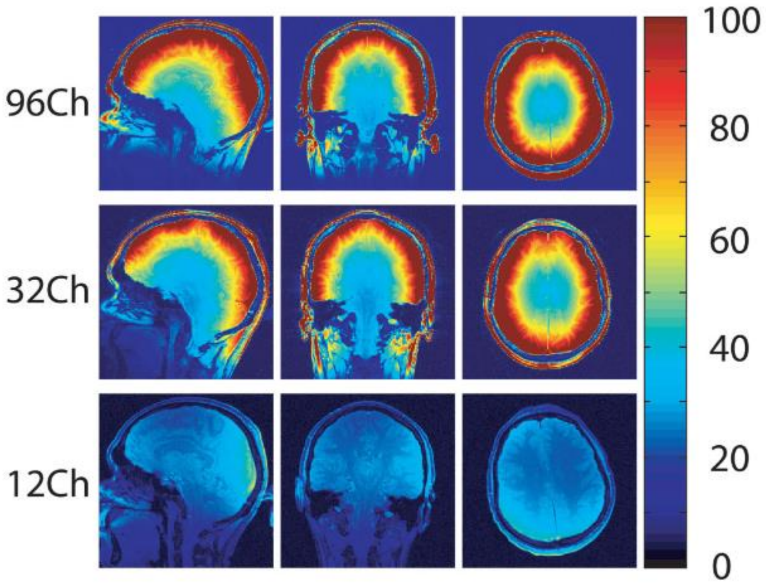
Roemer et al., MRM (1990)

MULTI-CHANNEL ARRAYS: MORE IS BETTER?

- We use arrays, because we need:
  - SNR of surface coil
  - FOV of volume coil
  - Parallel imaging: higher acceleration factors


Schmitt et al., MRM (2008)

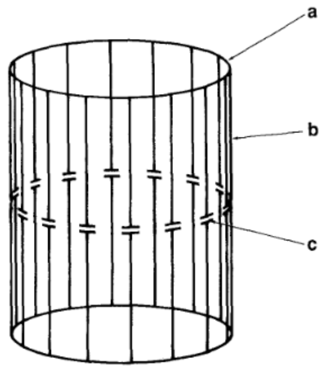

Wiggins et al., MRM (2009)

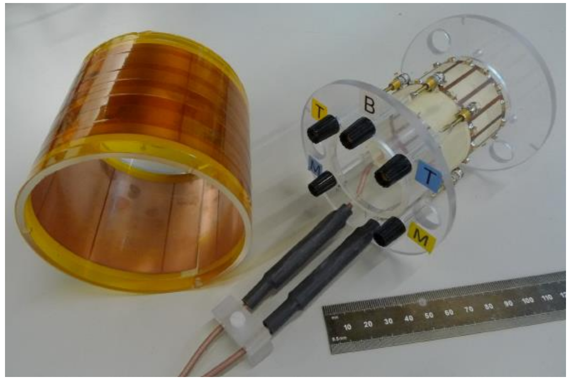


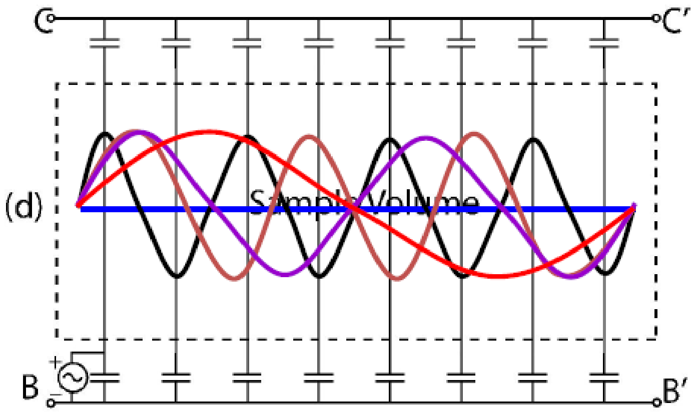
- More is not always better + MR system limitations + coupling

VOLUME COILS: BIRDCAGE COIL

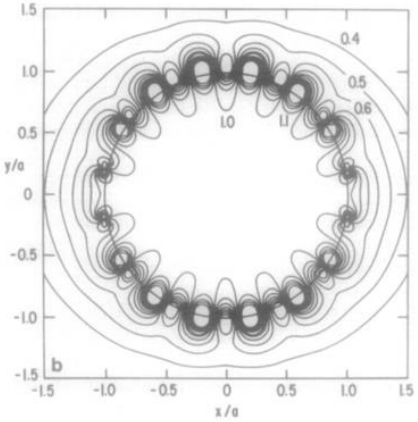
- Good approximation to sinusoidal current distribution
- One revolution of coil gives 2π phase shift
- Very high B_1 homogeneity
- Can be driven in quadrature
- Workhorse of clinical MRI







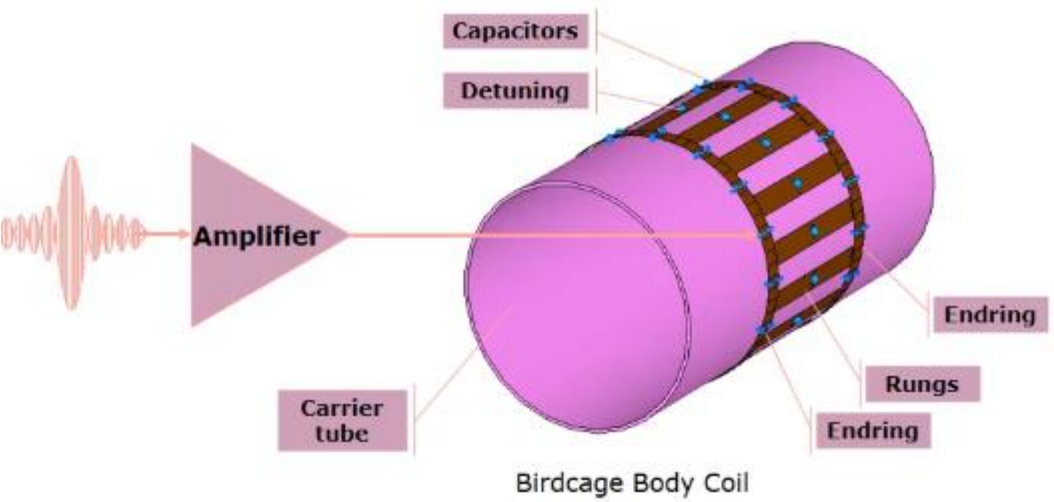
Modes 0, 1, 2, 3, 4



Hayes et al., JMR (1985)

TRANSMIT COIL – CLINICAL MRI

This is what you expect to see in clinical MRI machines: 1.5T and 3T



- **Homogeneous excitation** in a large volume
- **Peak power needed**:
  - 5 kW (<0.5T)
  - 15 kW (1.5T)
  - 35 kW (3T)

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ULTRA-HIGH FIELD MRI: CHALLENGES

Pros and cons of UHF-MRI

Table 1
A partial overview of potential pros and cons when increasing the magnetic field strength. Note that the consequences – pro or con – may depend on technical and anatomical details. Modified and expanded from [24].

	Characteristic	Trend as B_0 ↑	Pro	Con
	SNR	↑	Higher resolution, shorter scan time, X-nuclei feasible	None
*	SAR	↑	None	Fewer slices, smaller flip angle, longer TR, longer breathhold
	Physiological side-effects	↑	None	Dizziness, nausea, metallic taste
	Relaxation times	T1 ↑[a]	TOF, ASL, cardiac tagging	Longer scan time
		T2 ↓[b]		DWI, DTI
		$T2^*$ ↓	SWI, BOLD	
*	RF field uniformity	↓	Parallel reception Parallel transmission	Position-dependent flip angle, poor inversion, unexpected contrast
	Susceptibility effects	↑	BOLD, SWI, $T2^*$	Geometric distortions, intravoxel dephasing
	Chemical shift	↑	Fat saturation, CEST, MR spectroscopy	Fat/water and metabolite misregistration

[a] Although for most applications T1 increases with B_0, an increasing contribution from chemical shift anisotropy can also result in a decrease in T1 relaxation times (e.g. in ^{31}P MRS; cf. Section 6.1).
[b] Although for most applications T2 decreases with B_0, for quadrupolar nuclei T2 can also increase with field strength (cf. Section 6.1).

Ladd et al., Prog Nucl Magn Reson Spectrosc (2018)

SPECIFIC ABSORPTION RATE

- Energy absorbed by sample dissipates as heat
- Measured using SAR:

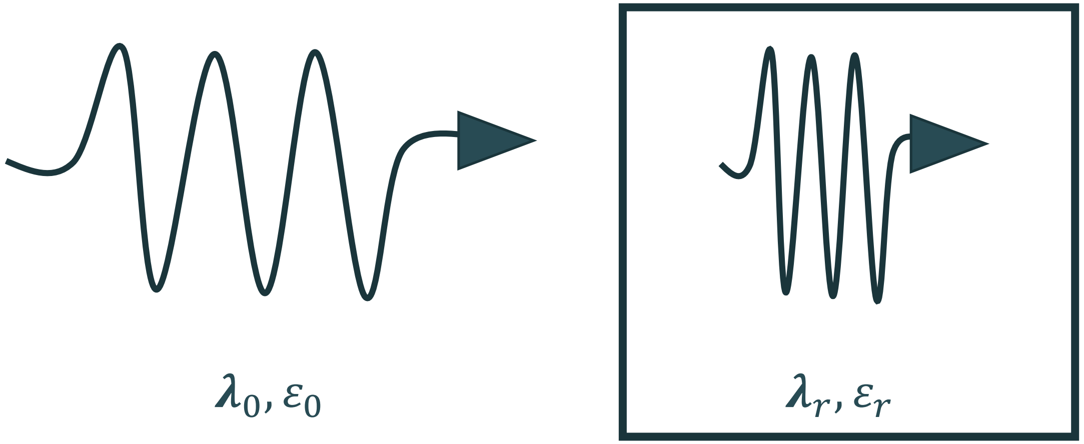
$$SAR = \frac{total\ RF\ energy\ dissipated\ in\ sample}{exposure\ time\ \times\ sample\ weight}$$

- Human body is heterogeneous and SAR is spatially-dependent $(\vec{r})$:

$$SAR = \frac{1}{V}\int \frac{\sigma(\vec{r})\left|\vec{E}(\vec{r})\right|^2}{\rho(\vec{r})} d\vec{r}$$

σ – tissue conductivity
ρ – tissue density
V – tissue volume

WAVELENGTH VS. MAGNETIC FIELD STRENGTH



$$\lambda_r = \frac{\lambda_0}{\sqrt{\varepsilon_r}}$$

- λ_0 - wavelength in the air
- ε_r - dielectric permittivity

Magnetic field strength in [T]	Wavelength in brain [cm]
1.5	52.0
3.0	30.0
7.0	14.0
9.4	10.5
10.5	9.5
11.7	8.6
14.0	…
20.0	…

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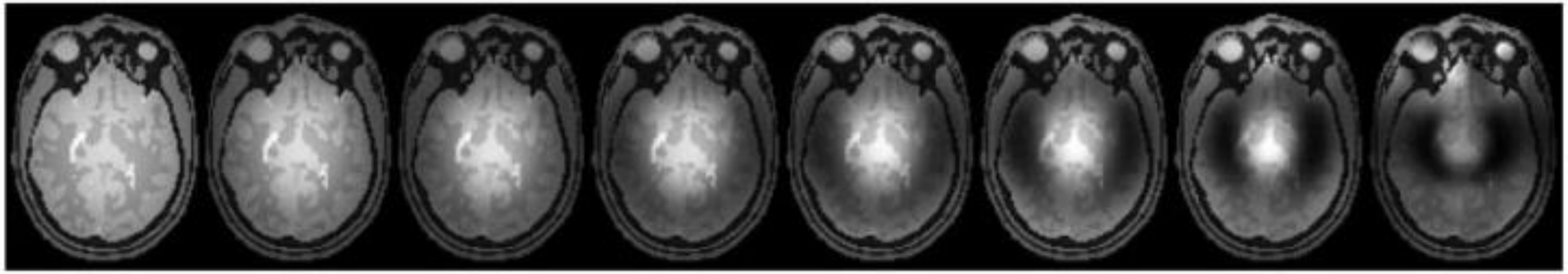
B_1 INHOMOGENEITIES

Birdcage coil



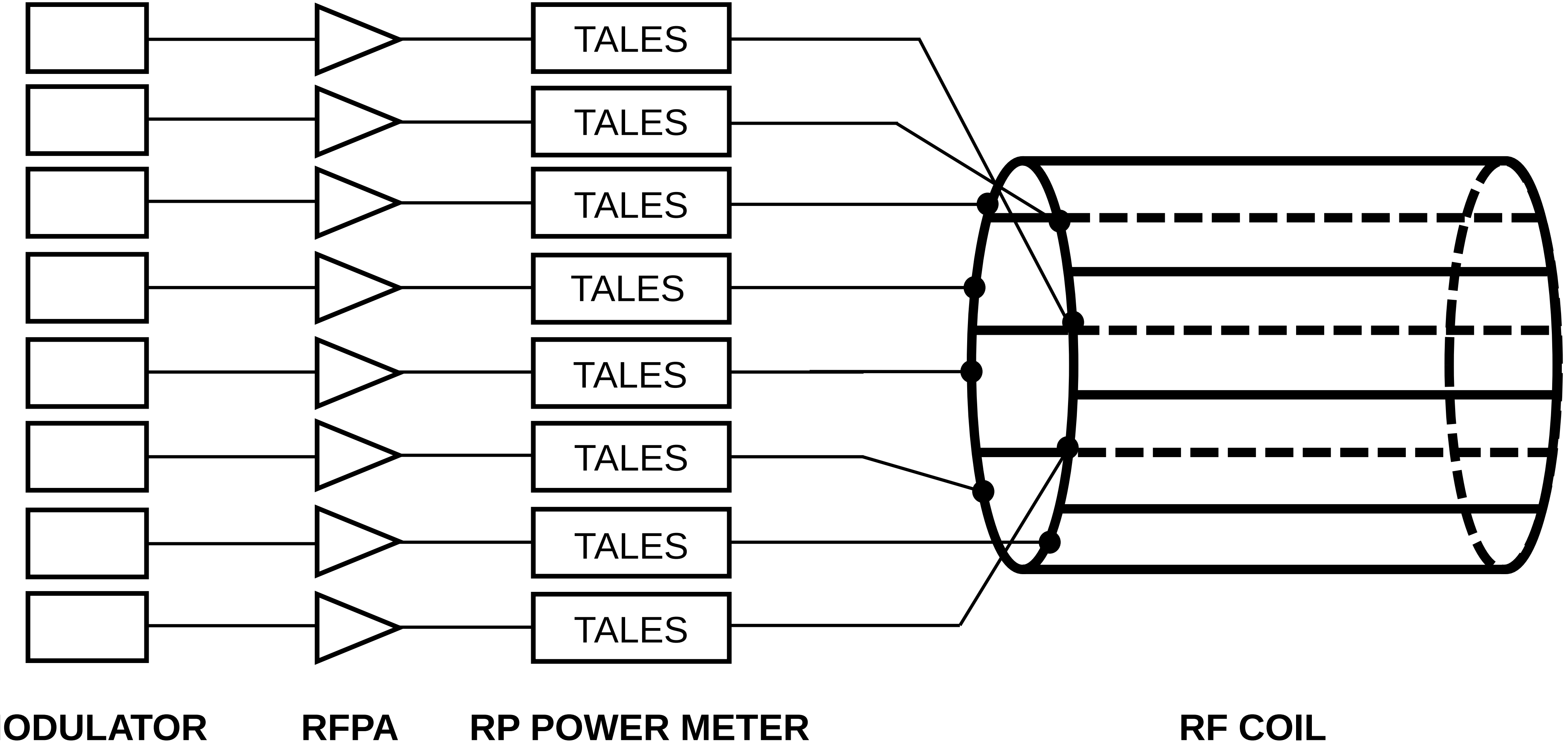
- Q: would it be the same for a preclinical 9.4T and 14T MRI scanner?

Webb and Collins, IMA, (2010)

PARALLEL TRANSMIT (PTX) SYSTEM

Body coils:
- 5 kW (<0.5T)
- 15 kW (1.5T)
- 35 kW (3T)

Local TXRX:
- **8 kW (7T)**



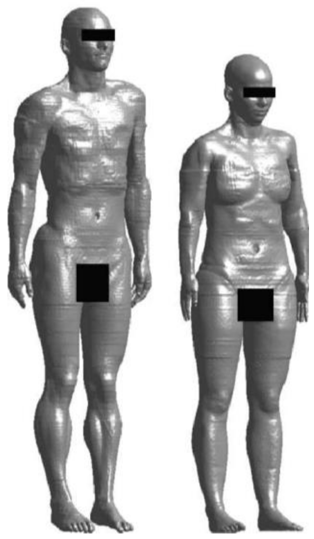
RF SHIMMING

AMPLITUDE CONTROL

$$\left|\sum_{i}^{N}\left|B_{1i}^{+}(\vec{x})\right|\right| \rightarrow \begin{matrix} B_{1coil_1}^{+}(\vec{x}) \\ \vdots \\ B_{1coil_N}^{+}(\vec{x}) \end{matrix} \rightarrow \begin{matrix} \alpha_1 B_{1coil_1}^{+}(\vec{x})\, e^{-j\omega t+\varphi_1} \\ \vdots \\ \alpha_N B_{1coil_N}^{+}(\vec{x})\, e^{-j\omega t+\varphi_N} \end{matrix}$$

PHASE CONTROL

HUMAN VOXEL MODELS:



B_1^+ HOMOGENEITY — SAR — B_1^+ EFFICIENCY

$$f(\Phi) = \frac{std(|B_1^+(\Phi)|)}{mean(|B_1^+(\Phi)|)} - \beta * \frac{1}{\sqrt{max\left(SAR_{10g}^{Duke}(\Phi),\ SAR_{10g}^{Ella}(\Phi)\right)}} * \frac{MOS(|B_1^+(\Phi)|)}{SOM(|B_1^+(\Phi)|)}$$

$$MOS(|B_1^+(\Phi)|) = \left|\sum_i B_1^+(\Phi_i)\right| \quad , \quad SOM(|B_1^+(\Phi)|) = \sum_i |B_1^+(\Phi_i)|$$

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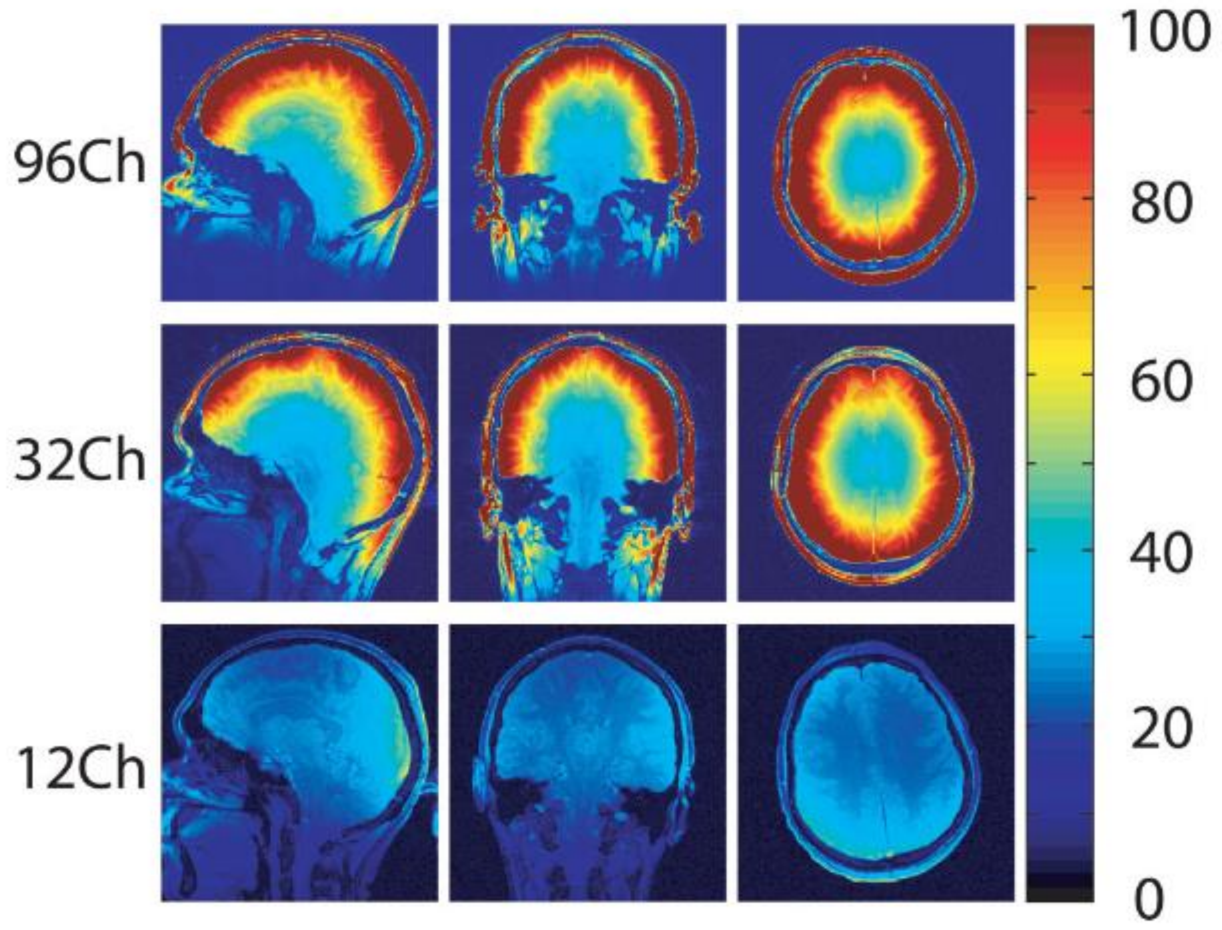
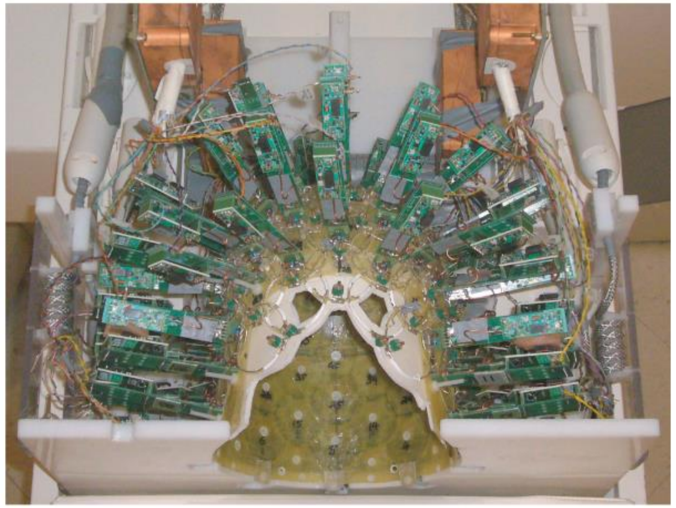
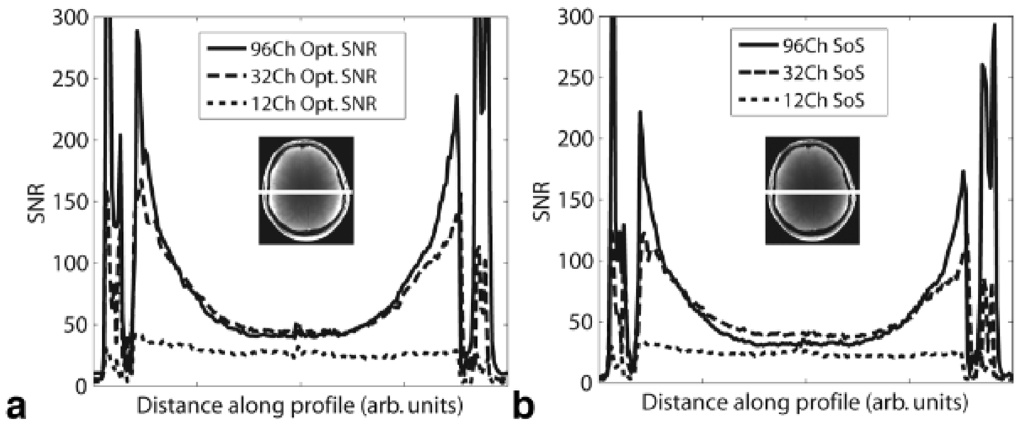
MORE IS BETTER? LESSONS FROM 3T







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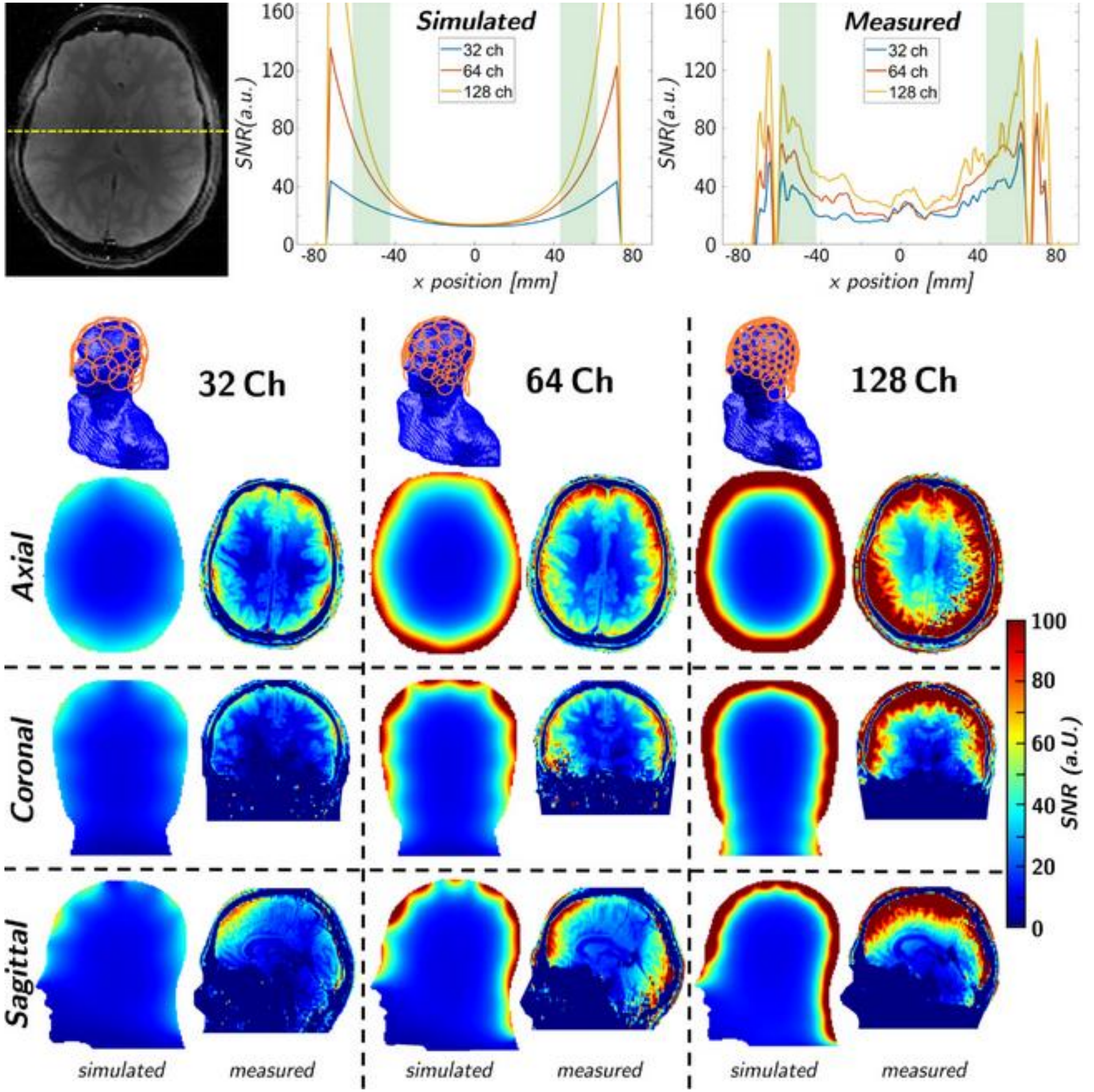
Wiggins et al. MRM (2009)

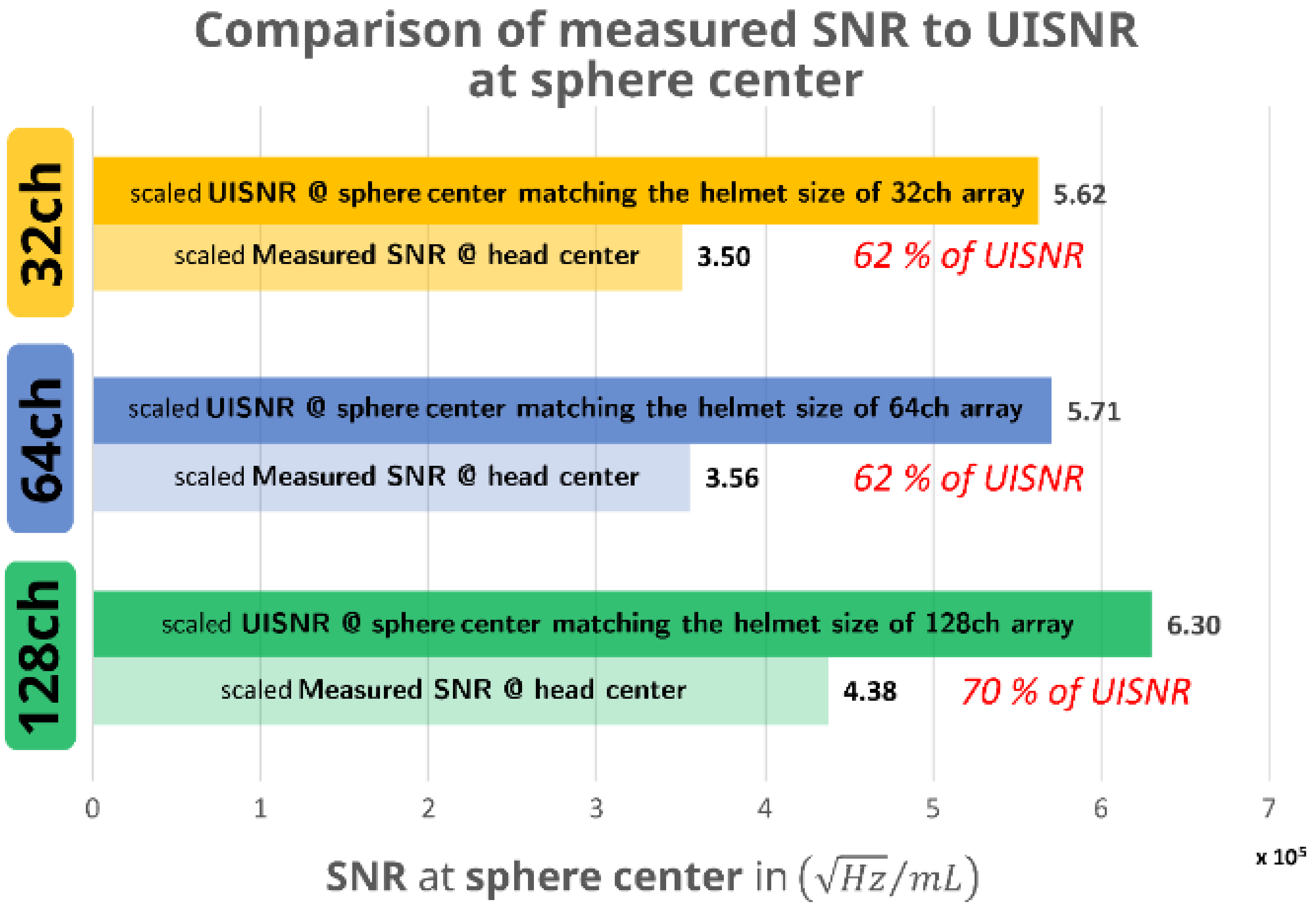
MORE IS BETTER @7T?

128-channel receive array



Gruber et al. MRM (2023)

MORE CHANNELS @ 7T VS. UISNR



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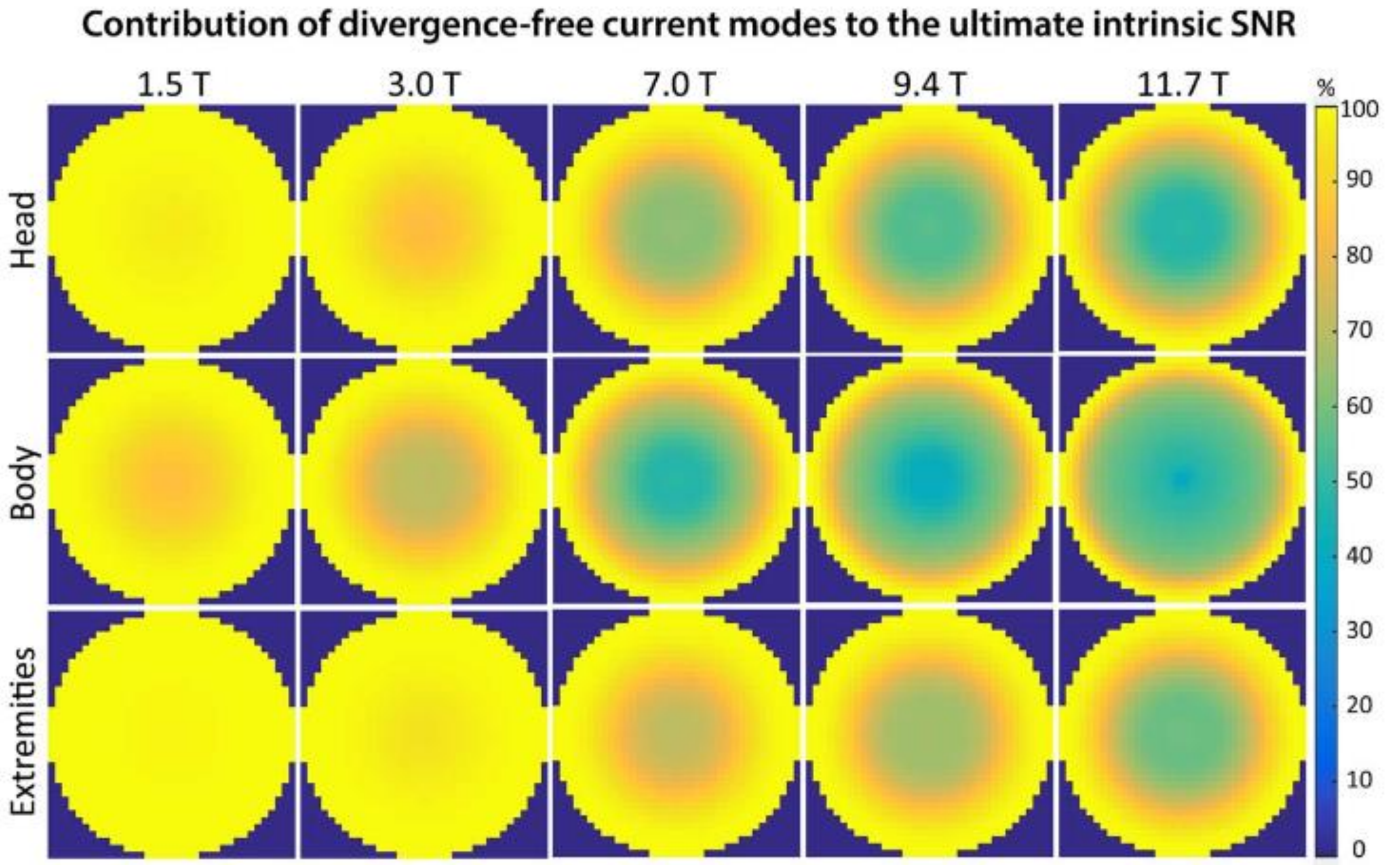
COMBINATION OF LOOPS AND DIPOLES



Lattanzi et al., MRM (2018)

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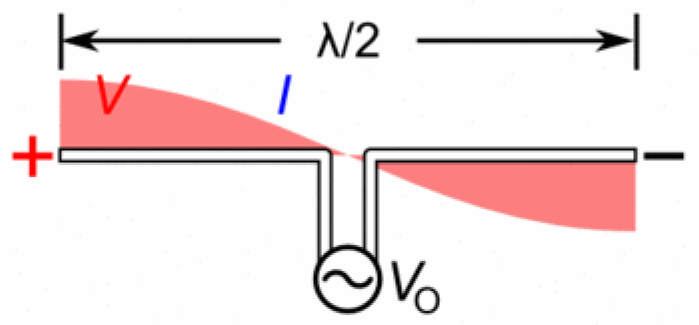
DIPOLE ANTENNAS



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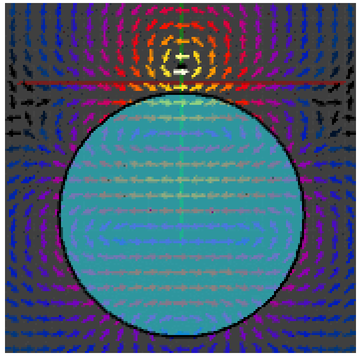
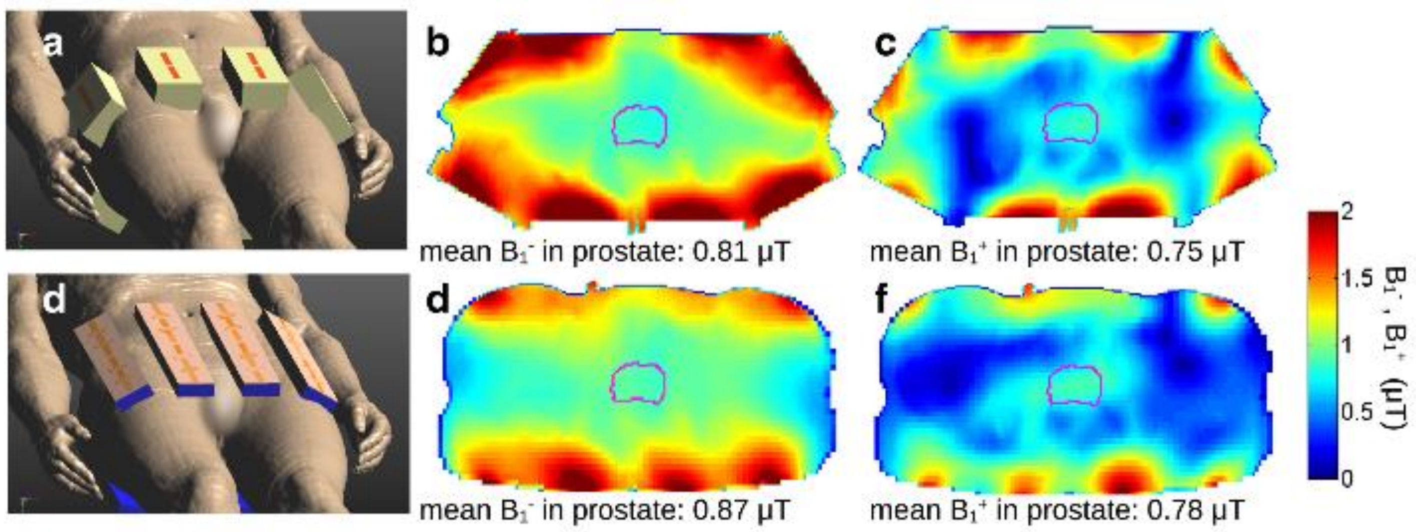
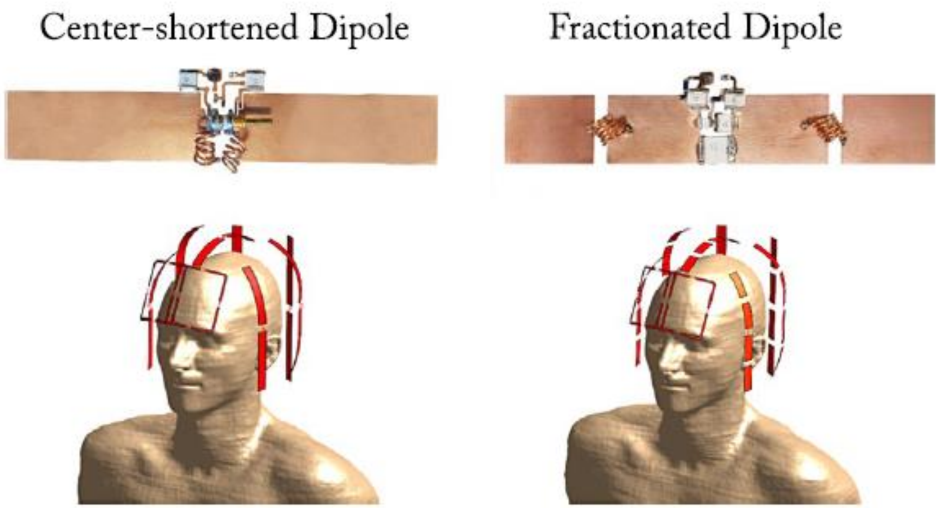




Raaijmakers et al., MRM, (2016)



Clement et al., MRM, (2019)

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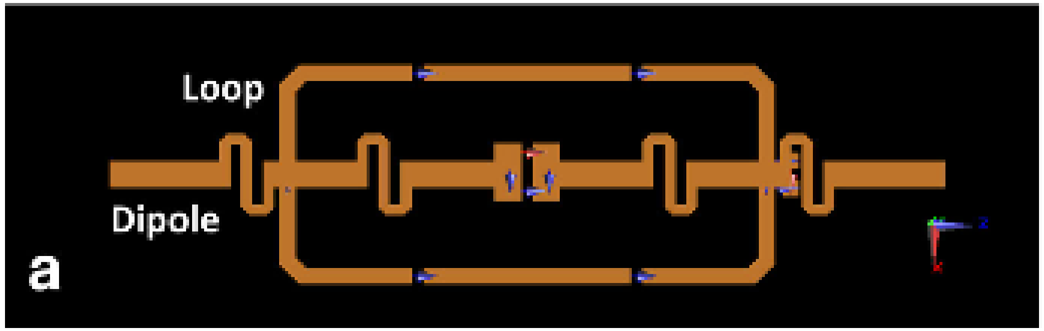
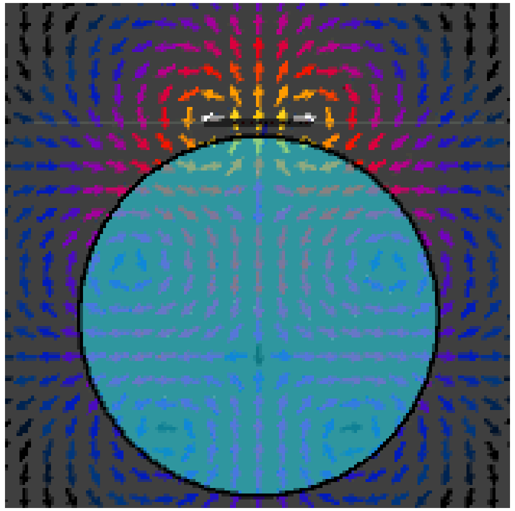
COMBINATION OF LOOPS AND DIPOLES

H-Field



Wiggins et al., Proc Intl Soc Mag Reson Med (2013)   Erturk et al., MRM (2017)

LOOP



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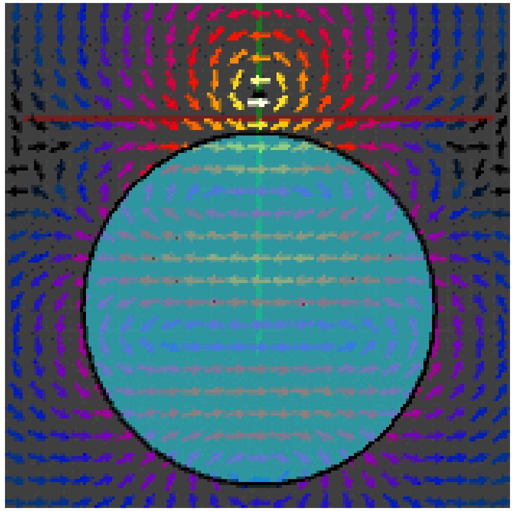
DIPOLE



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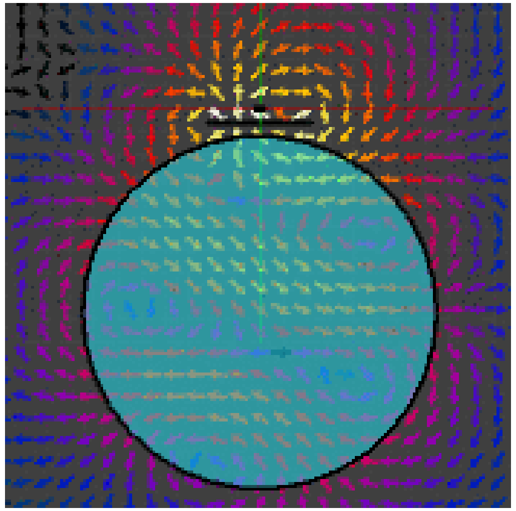
LOOP/DIPOLE



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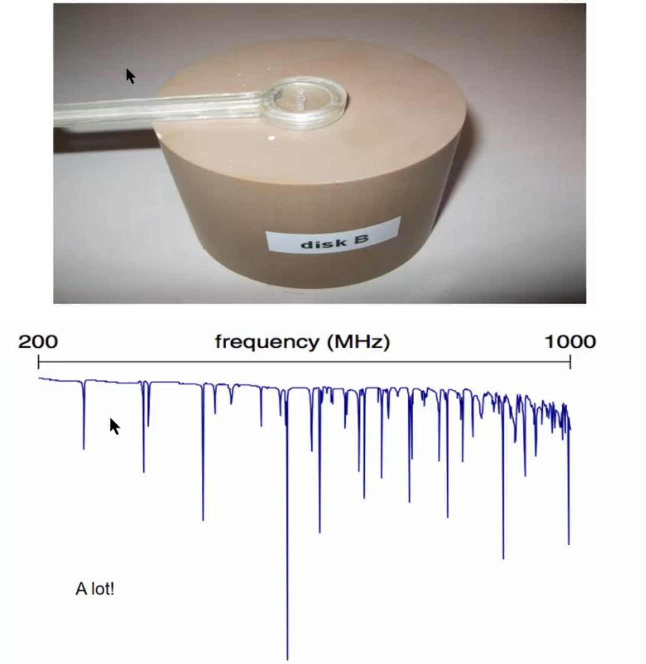
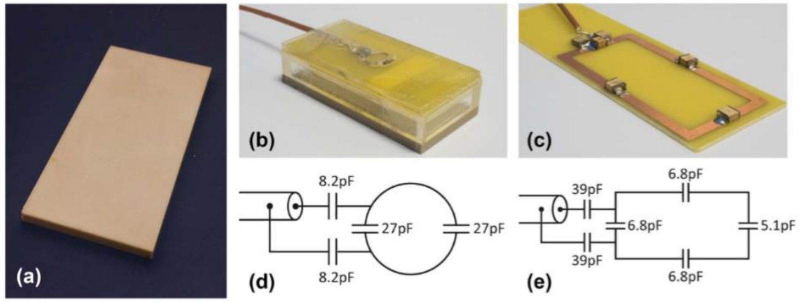
DIELECTRIC RESONATORS

Dielectric modes: stable, time-invariant electric- and magnetic field patterns formed within the resonator



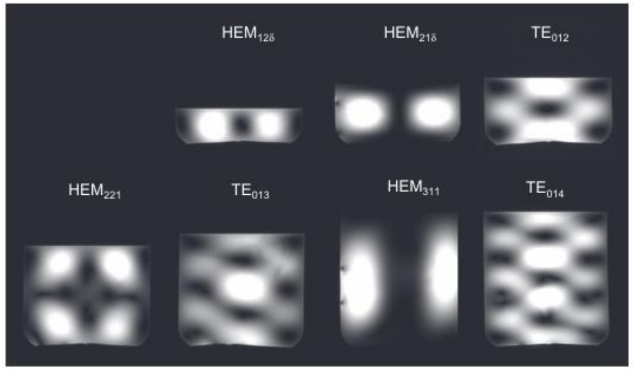
S. Aussenhofer, "Dielectric Materials & Resonators", Educational Course ISMRM Annual Meeting 2016



T. O'Reilly et al, MRM (2017)

S. Aussenhofer, et al, JMR (2014)

DR modes visualization at 7T



A.G, Webb, JMR (2012)

THANK YOU FOR YOUR ATTENTION