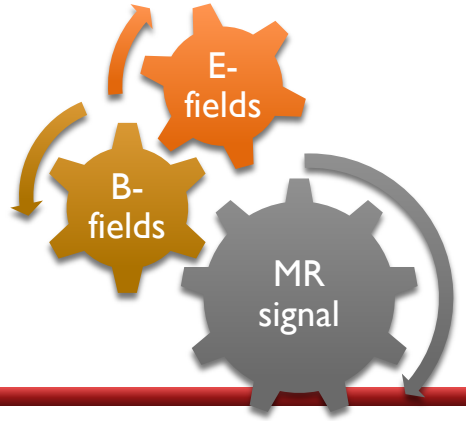
IEEE-CNSV

Consultants' Network of Silicon Valley



HOW ULTRA HIGH-FIELD MRI IS REVOLUTIONIZING MEDICINE

Open Engineering and Multiphysics Challenges

SIMONE A WINKLER, PHD, PE

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ASSISTANT PROFESSOR, WEILL CORNELL MEDICINE

INGENUYX
ENGINEERING SOLUTIONS

Weill Cornell Medicine
UHF MRI Program

OUTLINE

- NIH funded Motivation: Human Connectome Project using UHF MRI
- Introduction to UHF MRI
- Engineering-related challenges in UHF MRI
  - Image shading due to B1 inhomogeneities
  - Image distortion and artefacts due to B0 inhomogeneities
  - MR Safety
  - MR Acoustics

MR BASICS – NUCLEAR MAGNETIC RESONANCE

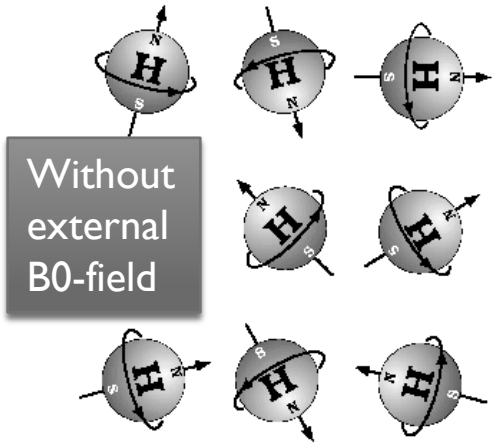
- Isotopes in the body with odd number of protons possess nonzero spin → moving electrons generate small magnetic field
- Hydrogen atoms most abundant in human body (tissue water content)

TWO-STEP PROCEDURE:

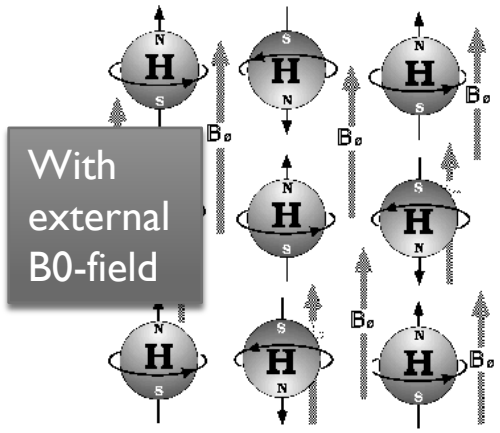
1. Alignment (polarization) of spins with an external static magnetic field B0
2. Perturbation of alignment by use of an RF (electromagnetic) field → magnetization of protons can be measured

MR BASICS – STEP 1: SPIN ALIGNMENT

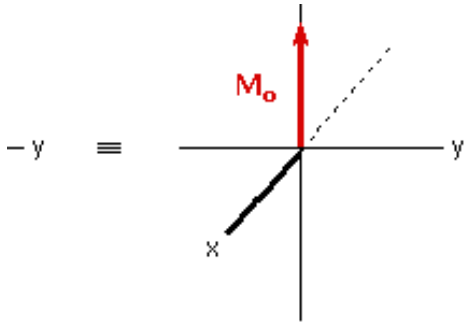
Alignment



No net magnetization



A slight excess number of protons align with Bø.

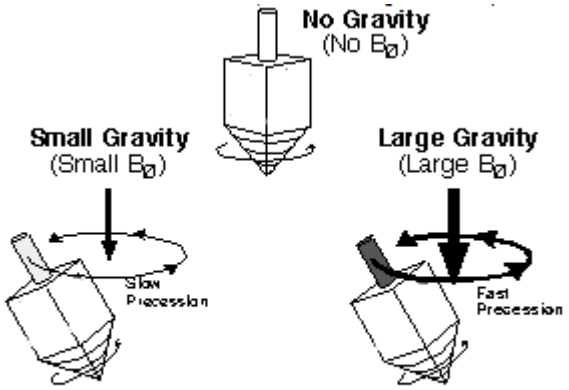


M_o = Net Macroscopic Magnetizatior

Precession

Courtesy B. Hargreaves

Spin precession at Larmor frequency



$$\omega_{ø} = \gamma B_{ø}$$

Larmor frequency depends on magnetic properties of proton, and is proportional to static magnetic field strength

MR BASICS – STEP 2: PERTURBATION/MEASUREMENT

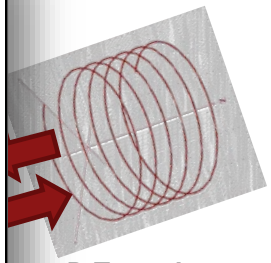
Idea: measure magnetic field that spin generates!

excitation

Precession at Larmor Frequency

Transverse component can be detected with an RF coil!

RF excitation:
Additive B1 pulse in *transverse* plane at Larmor frequency



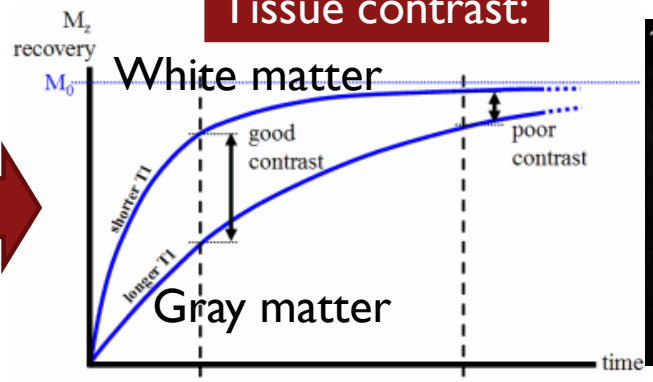
RF coil transceives transverse magnetic field

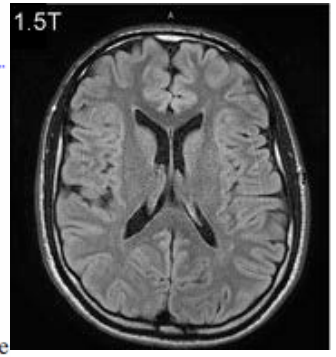
relaxation

After RF pulse is turned off:

3T

Tissue contrast:

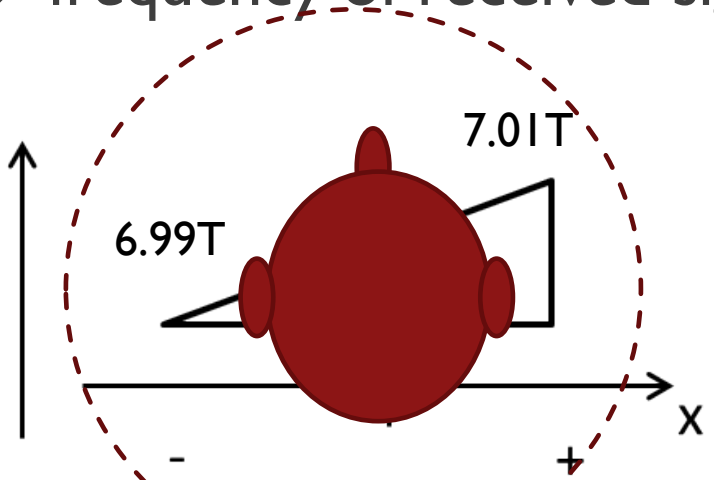




white matter: T1=832ms, T2=110ms
gray matter: T1=1331ms, T2=80ms

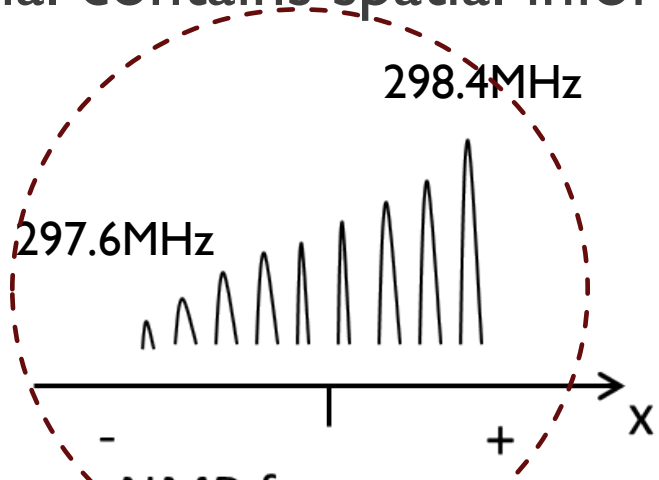
MR BASICS – SPATIAL LOCALIZATION

- Larmor frequency depends on static magnetic field strength
- Small magnetic field variation is overlaid depending on location of imaged voxel
- → frequency of received signal contains spatial information!



Field gradient G_x

$B_Z = B_0 + G_x X$

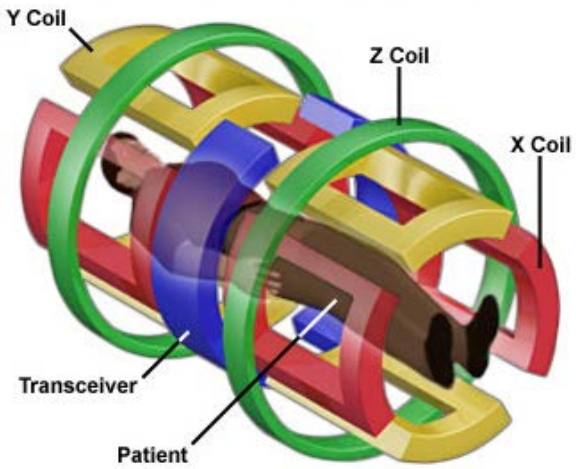


NMR frequency

$\nu(x) = \nu_0 + \gamma G_x X/h$

With **linear** additive gradient: Spatial information can be recovered from Inverse Fourier Transform!!

3D localization



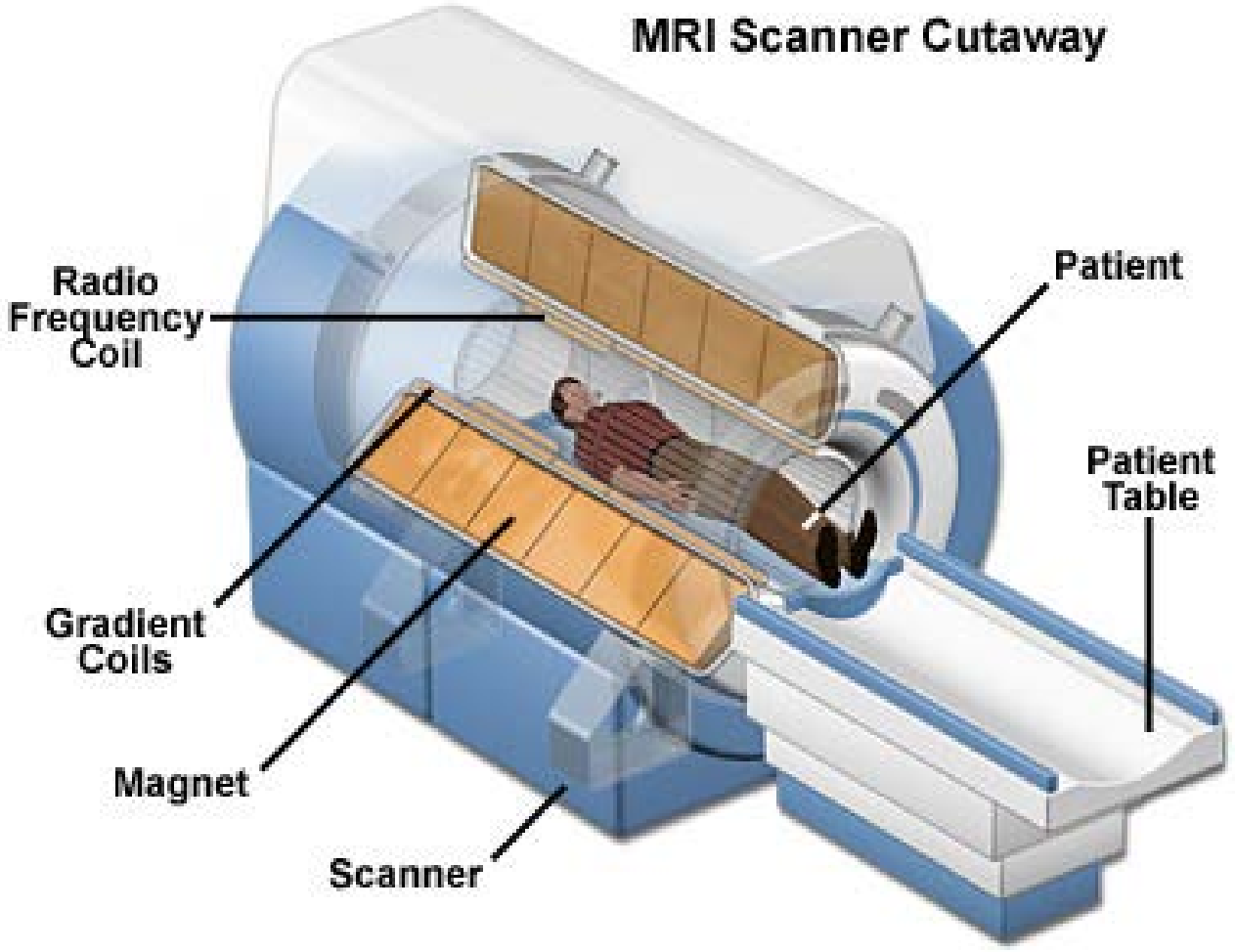
Gradient coils DC current generates linear B0 variation

MR BASICS - SYSTEM

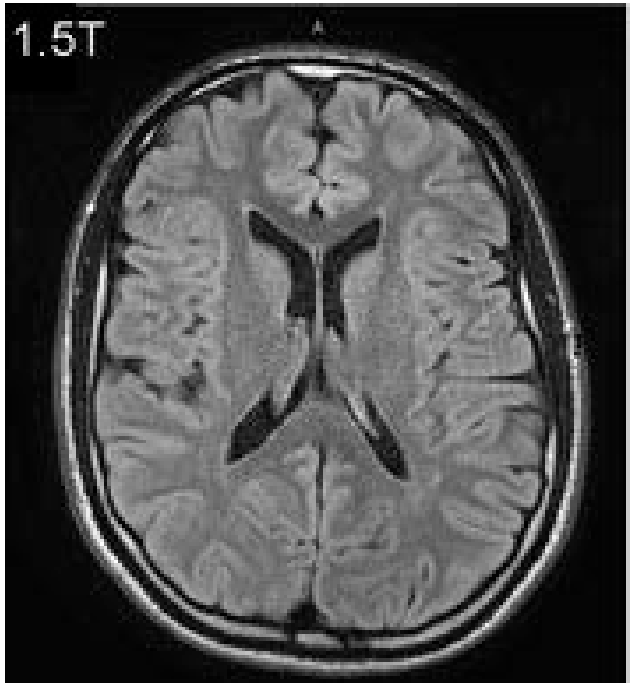
Main system components:

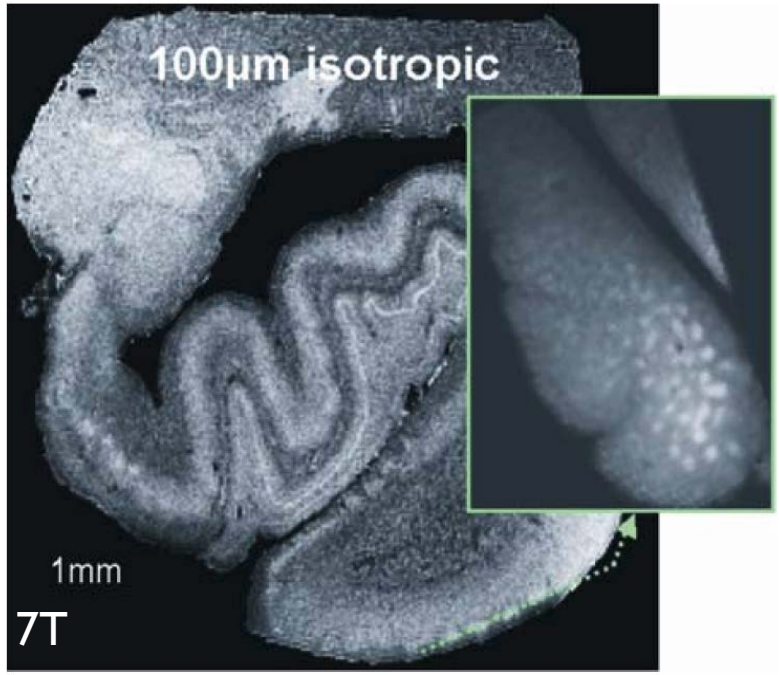
1. Magnet (generates B0 field → aligns spins → net magnetization)
2. RF coil (excites and receives RF magnetic field at Larmor frequency)
3. Gradient coils (localization by small variation of B0 field)

Typical B0 field strengths (human):
Clinical: 1.5T-3T
Research: 7T-11.7T

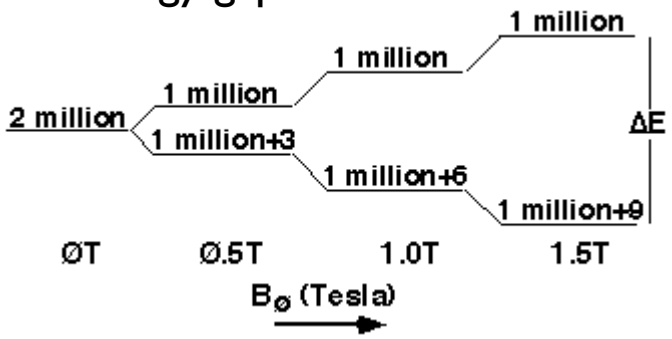


WHY UHF MRI?





Energy gap:



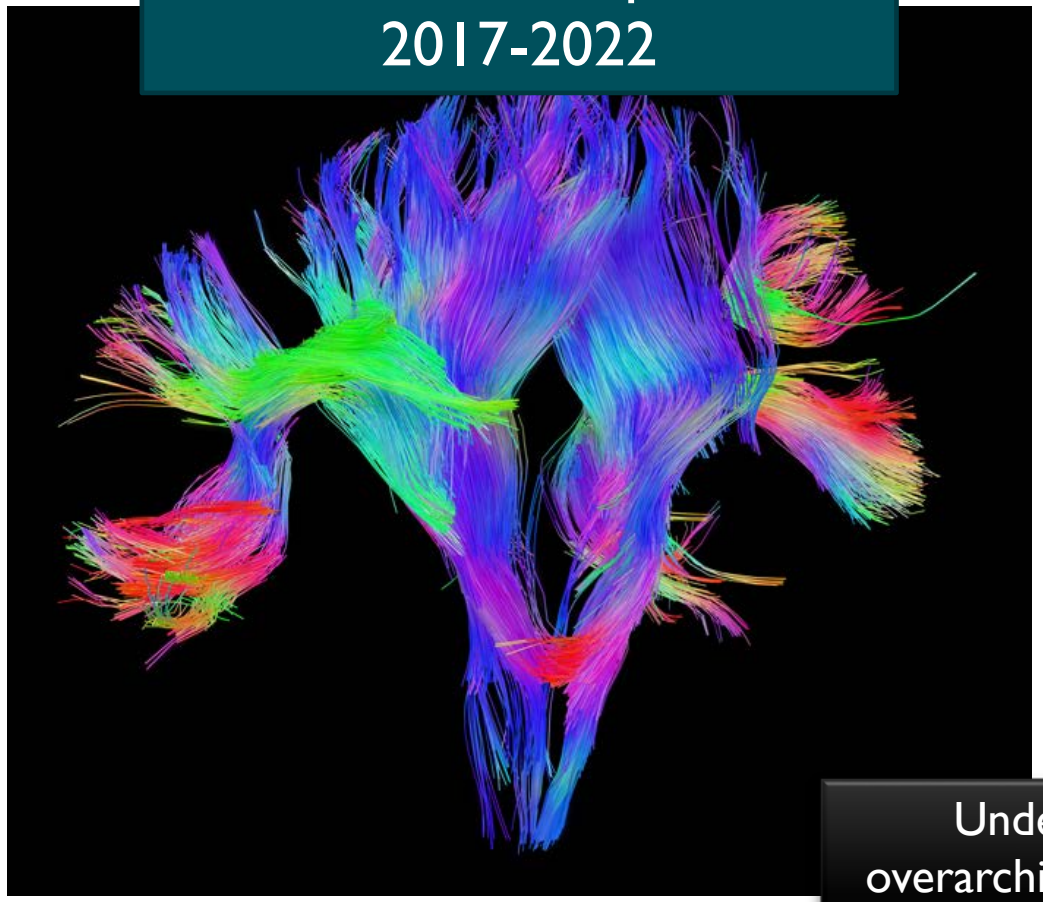
- Increased sensitivity for proton imaging applications (SNR increases approx. proportionally with B0)
- Allows for anatomical imaging with higher spatial resolution
- Use higher sensitivity to examine dynamic and functional aspects (BOLD, other brain activity)

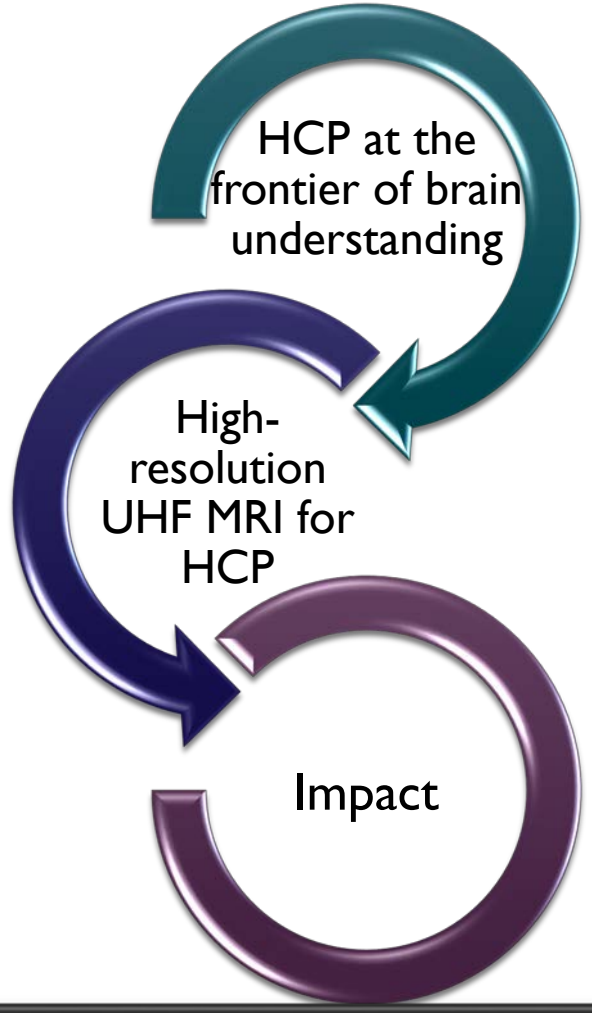
- Potential for detection of new physiology of healthy and diseased tissue
- New forms of contrast

HIGH-END MR INSTRUMENTATION FOR HCP

Human Connectome Project

NIH K99/R00 funding
Transition to Independence
2017-2022





- Map neural connectivity pathways in vivo → "brain circuitry map"
- Part of Obama's BRAIN initiative
- Individual patient connectome (IPC) as future clinical diagnostics tool

- Next generation UHF HCP
- Overcoming critical UHF barriers
- Long-term clinical potential of UHF MRI
- Improved hardware for 3T MRI

Understanding of neuropsychiatric disorders and overarching other neurological/neurodegenerative disease

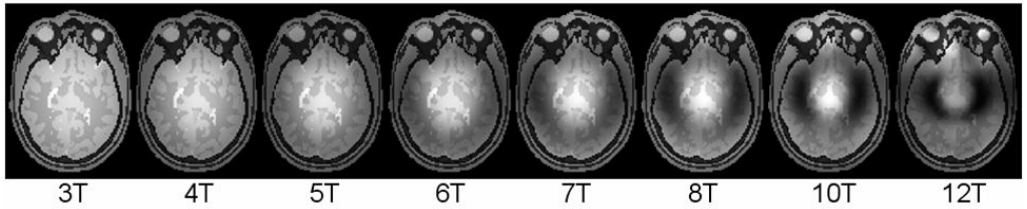
HIGH-FIELD MRI AT STANFORD

- GE Discovery MR950 7T scanner
- 2-ch Tx / 32-ch receive modality
- 8-ch pTx / 32-ch receive modality

Not FDA approved

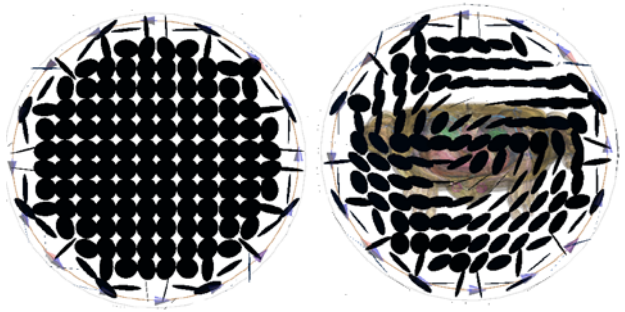
Approximately 50-100 7T scanners throughout the world only!

HARDWARE RELATED UHF MRI CHALLENGES



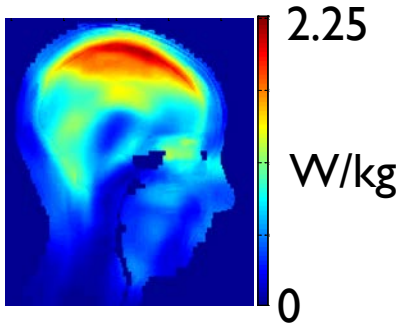
128MHz ····· Larmor frequency ····· 512MHz

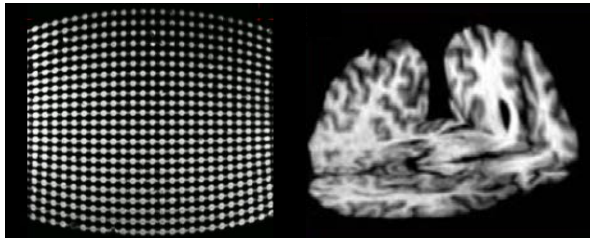
- Global SAR increases with B_0^2
- Local SAR becomes more inhomogeneous due to wave effects
- Local SAR is strongly position- and coil-dependent



B1 inhomogeneities → Image shading

RF heating in tissue (SAR) → Patient safety





B0 inhomogeneities → Geometric distortion & artefacts

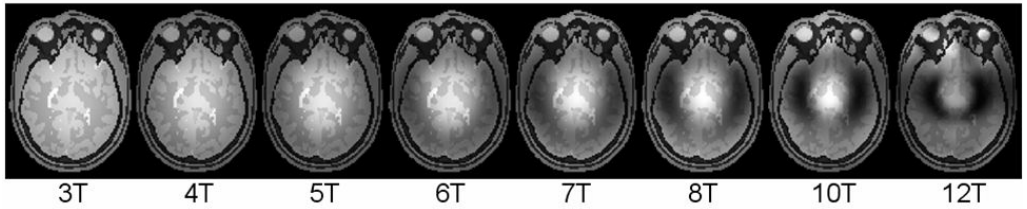
High gradient currents → Acoustics/Vibrations & cooling requiments

- Spatial encoding depends on B0 homogeneity
- Signal loss

- Lorentz forces increase with B_0
- Higher sound pressure levels
- Stronger vibrations

- Higher currents require dedicated cooling concepts

HARDWARE RELATED UHF MRI CHALLENGES



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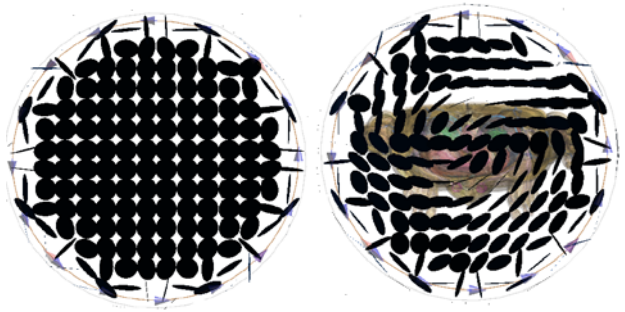

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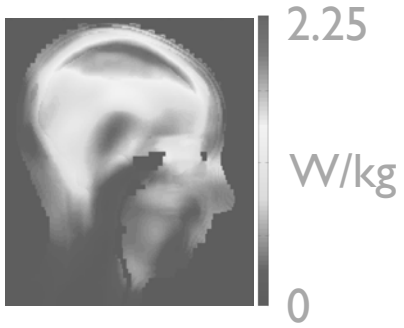

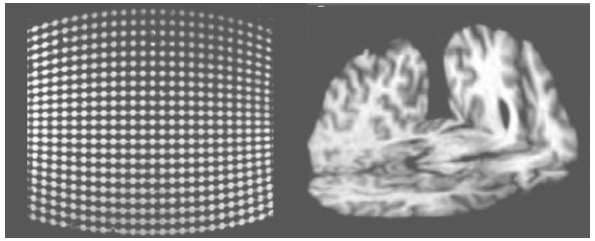

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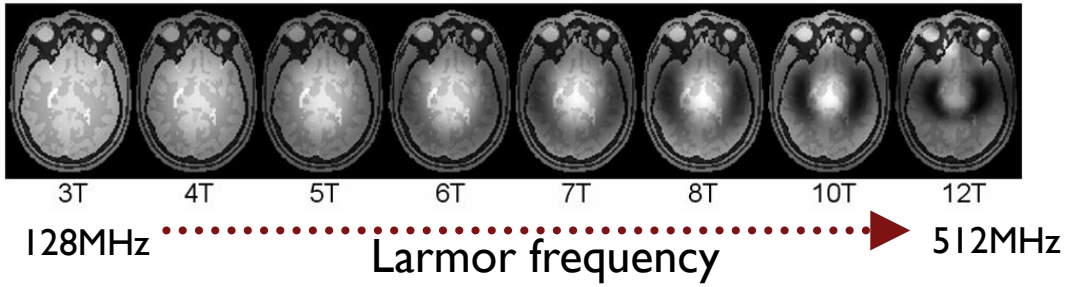
IMAGE SHADING – B1 INHOMOGENEITY

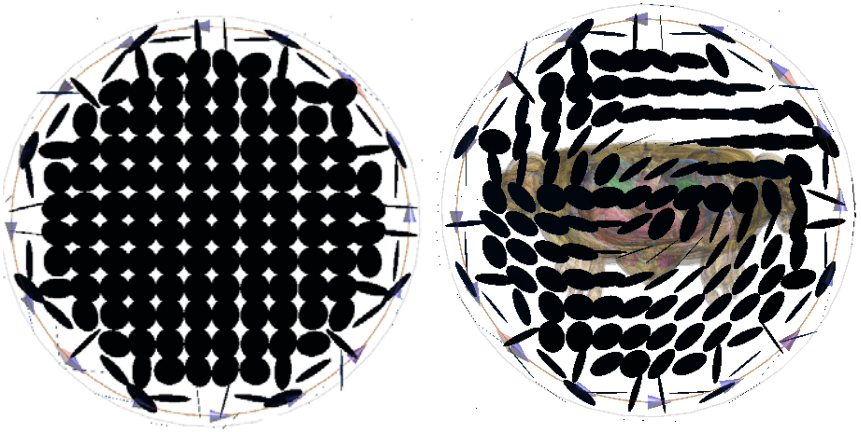
OPTIMIZED DIELECTRIC SHIMMING – SIMONE WINKLER, ALESSANDRO SBRIZZI, ET AL.

16-CHANNEL PTX COIL DESIGN – RICCARDO STARA, SIMONE WINKLER, ET AL.

B1 INHOMOGENEITY BASICS

Challenge:



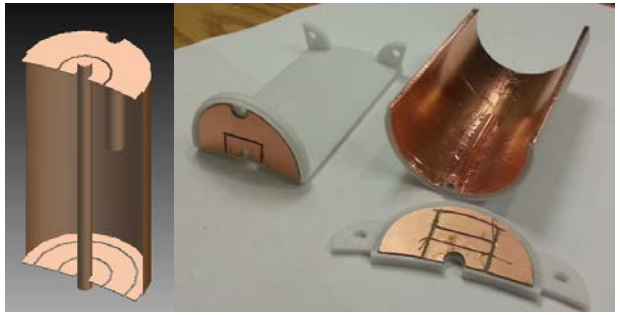


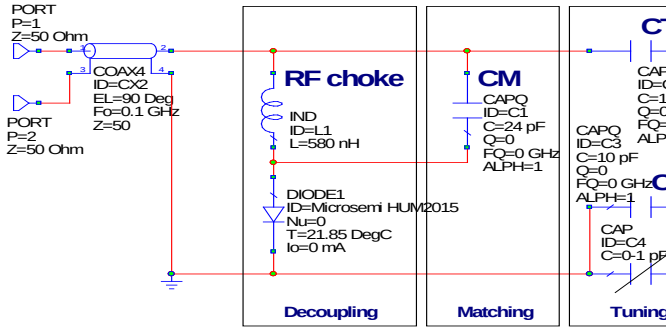
Remedy:

- Parallel Transmission (pTx): excitation from multiple independent channels with different pulse signal shapes at each channel
- RF shimming: excitation from multiple independent channels; only amplitude and phase are varied
- Dielectric shimming: passive perturbation of E- and B-fields by insertion of dielectric material that focuses field lines

16-CHANNEL PTX COIL DESIGN

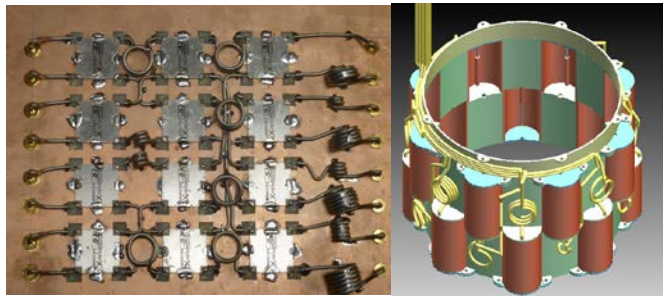
1. Novel coil element



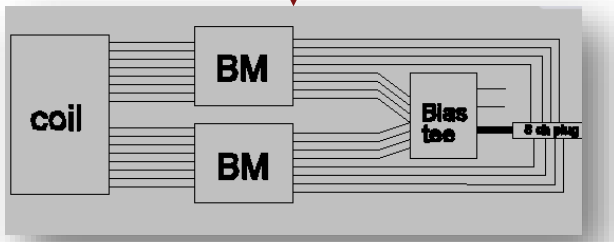


Inherent inter-element decoupling

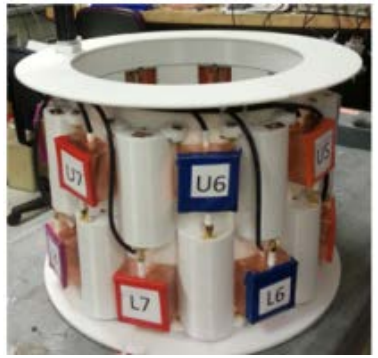
2. Modular configuration



1. 8ch single row
2. 8ch checkerboard
3. 16ch with two Butler matrices
4. 8ch with Butler matrix
5. 2ch with Butler matrix
6. etc.



3. 3D segmentation



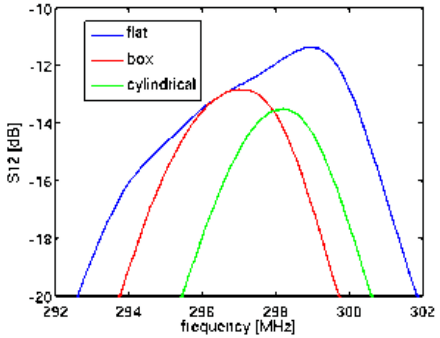
Z-segmentation

- Illumination of lower brain regions
- Better acceleration
- Better pTx flexibility

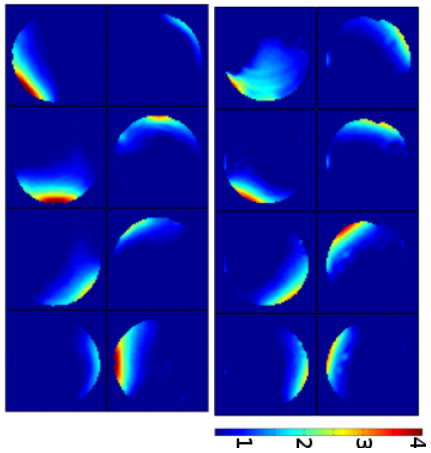
Stara et al., ISMRM 2014

Results

Inter-element coupling

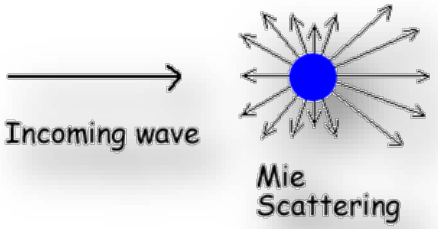


B1 maps

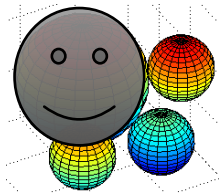


OPTIMIZED DIELECTRIC SHIMMING

- **Electromagnetic scattering is <u>ANALYTICALLY</u> described for dielectric-absorptive spheres!**
- **→ model dielectric pad as a sphere!**



- Model head as a dielectric sphere
- Model dielectric shimming pads as additional spheres
- Optimization to determine optimized position of the sphere(s)



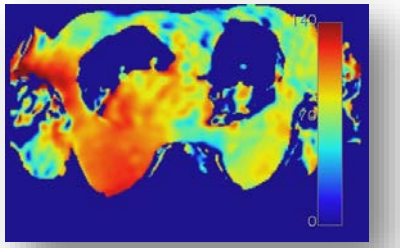
Requirement:

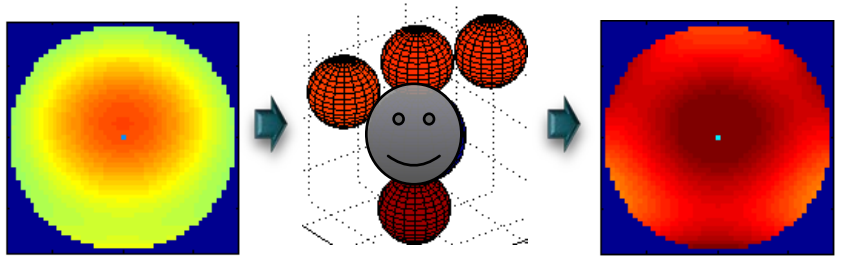
- Electromagnetic fields of unloaded RF coil expanded into <u>vector spherical harmonics</u>

$$\mathbf{H} = \frac{1}{Z_0}\sum_{l=0}^{\infty}\sum_{m=-l}^{l}\left[\boxed{p(l,m)}\boldsymbol{TM}_{lm} + \boxed{q(l,m)}\boldsymbol{TE}_{lm}\right]$$

$$\mathbf{E} = \sum_{l=0}^{\infty}\sum_{m=-l}^{l}\left[\boxed{q(l,m)}\boldsymbol{TM}_{lm} + \boxed{p(l,m)}\boldsymbol{TE}_{lm}\right],$$

Applications:
- 3T breast MR
- 7T neuroimaging



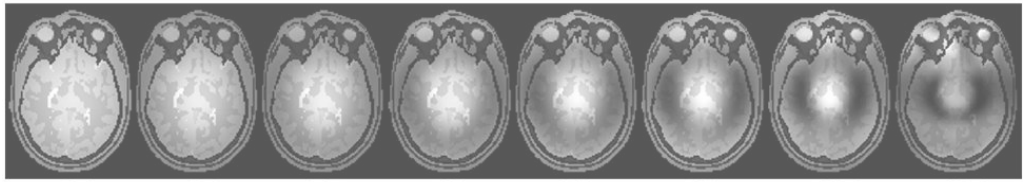


Improvement: 41.25%

Dielectric shimming also increases average B1+ field!

Winkler et al., Workshop Bremen 2014, Book Chapter 2014

HARDWARE RELATED UHF MRI CHALLENGES



128MHz ····· Larmor frequency ····· 512MHz

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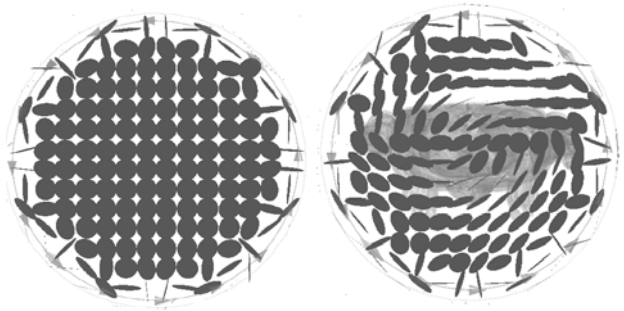
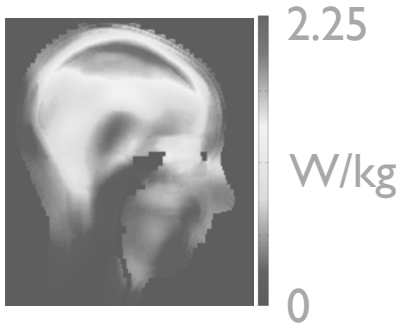
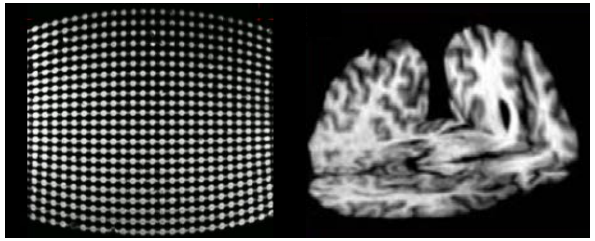


B1 inhomogeneities → Image shading

RF heating in tissue (SAR) → Patient safety





B0 inhomogeneities → Geometric distortion & artefacts

High gradient currents → Acoustics/Vibrations & cooling requiments

- Spatial encoding depends on B0 homogeneity
- Signal loss

- Lorentz forces increase with B_0
- Higher sound pressure levels
- Stronger vibrations

- Higher currents require dedicated cooling concepts

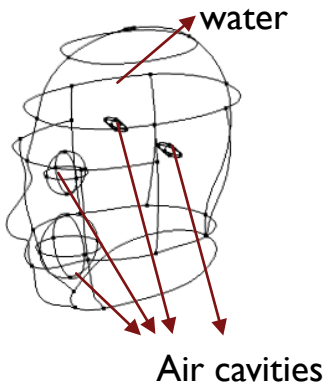
B0 INHOMOGENEITIES

RF SHIM COIL AT 7T – SIMONE WINKLER, JASON STOCKMANN, ET AL.

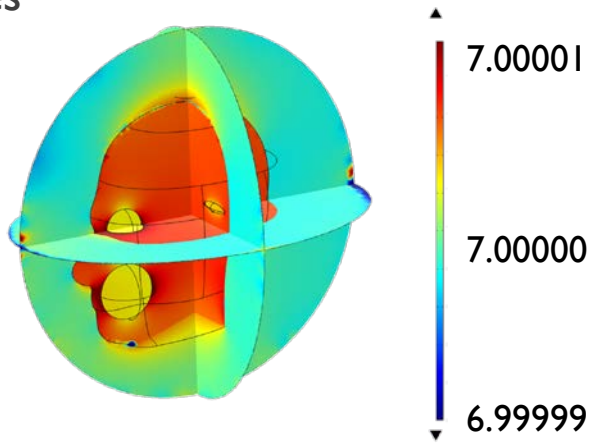
B0 INHOMOGENEITY BASICS

Challenge:

- B0 field has to be extremely homogeneous (order of ppm, for 7T in the 0.25µT range)
- Spatial encoding depends on B0 homogeneity
- Geometric distortions and/or signal loss



- Sources of B0 inhomogeneity:
  1. Magnet construction
  2. Magnetic susceptibility differences in tissues; in particular: air cavities (ears, sinuses, etc.)



Remedy:

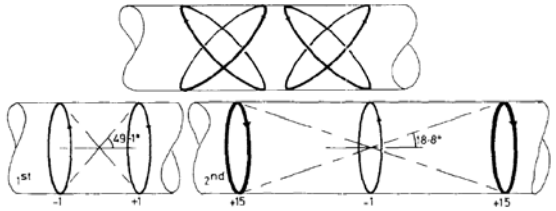
B0 shimming:

Passive

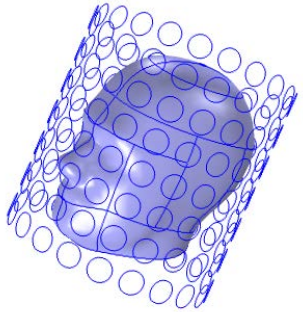
Iron inserts

Active

Spherical harmonics



Matrix shim



HISTORY

- High-order B_0 shimming is essential for modern MR neuroimaging
- Conventional approach: spherical harmonic shimming

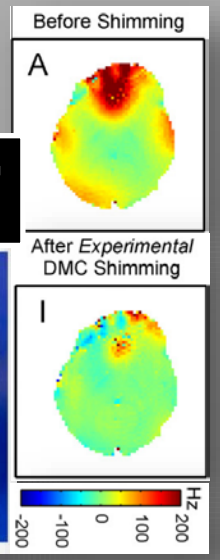
Multi-Coil Shimming

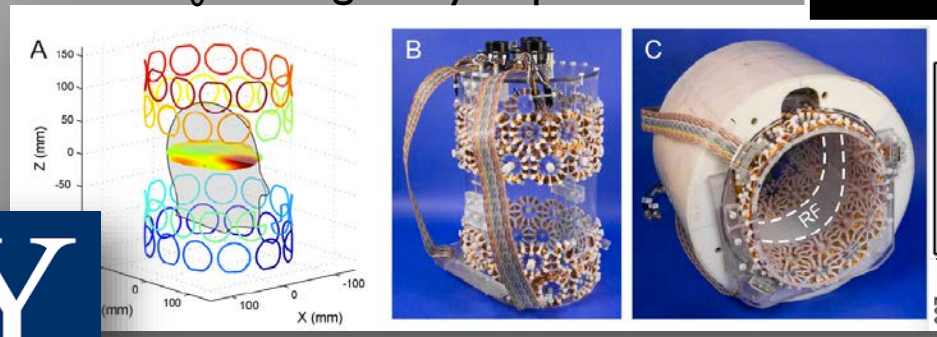
Alternative to spherical harmonic shimming

+ Efficient
+ Switch shim dynamically without eddy currents
- Restricted space for RF arrays
- Interactions with RF coils

Drastic B_0 homogeneity improvements

7T





Juchem C, JMR 212:280–288 (2011)

Multi-Coil + RF shimming

Uses RF chokes to bridge DC shim current into RF loop

+ Larger efficiency due to single-turn loop
+ Dynamic switching
- Construction complexity

3T

Truong TK, Neuroscience 103:235-240 (2014)

RF-SHIM HELMET - CONCEPT

RF-Shim Helmet - Concept:

3T

Combining Rx (RF) and B_0 (DC) shimming functions on the same conductor

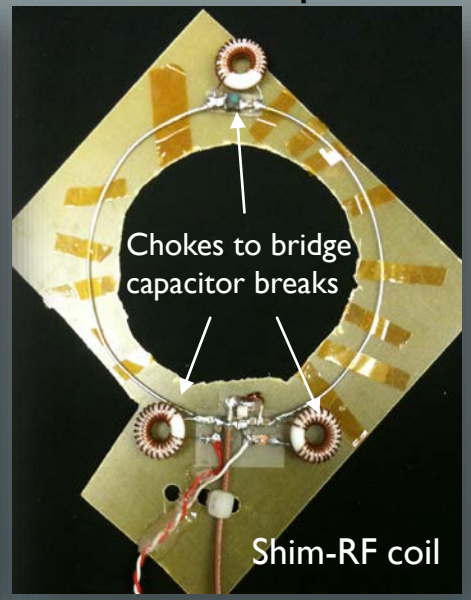
+

Array of single-turn loops
Helmet shape for closer proximity
(increased Rx SNR and B_0 shim efficiency)

Coil element

RF-path only

Added DC path

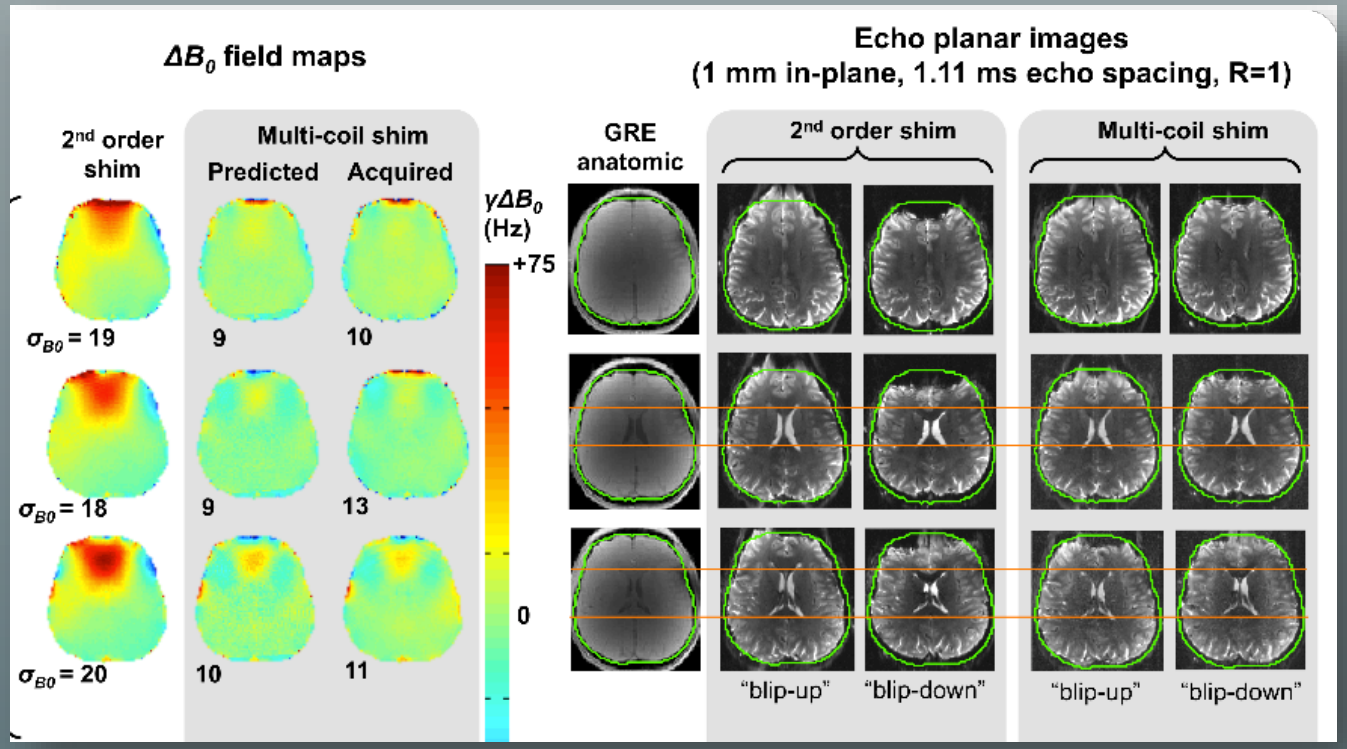


Stockmann, MRM early view (2015)

RF-SHIM HELMET - RESULTS

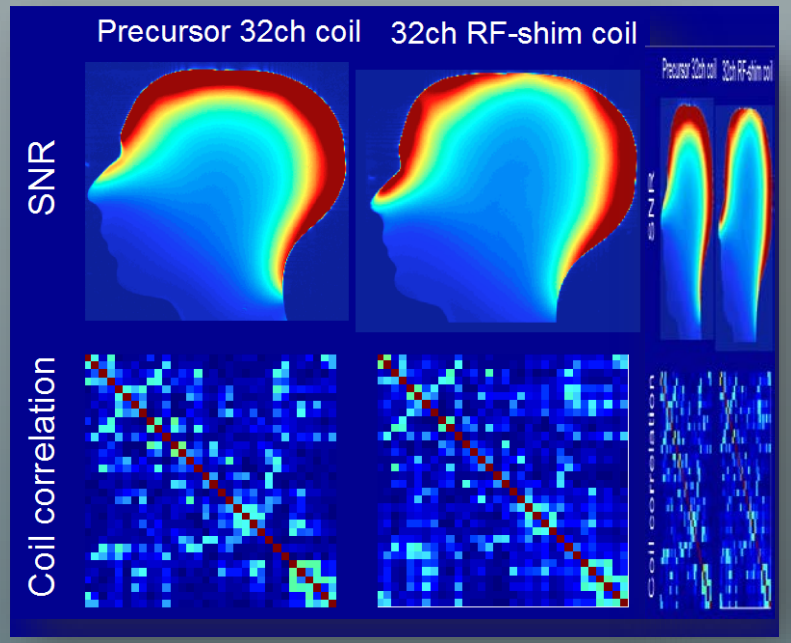
Rx-Shim Helmet – 3T Results:

B_0 – shimming results:



+ Reduced distortion and closer alignment for blip-up and blip-down
+ Predicted and measured field maps agree
+ Low current requiremnts (2.5 A max per coil)

RF - SNR results:



RF-shim coil exhibits modest SNR loss as compared to precursor RF-only coil

Stockmann, MRM early view (2015)

MY ROLE: RF-SHIM CONCEPT AT 7T

Using the RF-shim concept at 7T appears compelling:

1. Increased in-vivo B_0 distortion at 7T → greater impact for
   - EPI (fMRI, Diffusion)
   - QSM
   - Inversion pulses
2. SNR impact at 7T might be lower
   - A well designed receive array is body noise dominated
   - We hypothesize that at 7T added copper noise will have less impact

However:

- Conversion from 3T RF-only to RF-shim array caused 10-15% SNR loss
- We seek to minimize SNR loss at 7T

→ Optimized loop design becomes **essential**

The resulting 7T RF-shim array would offer:

- Helmet design for SNR and shim efficiency
- Minimal interference of RF and shim functions

→ Optimized compact tool for modern high-field MR neuroimaging

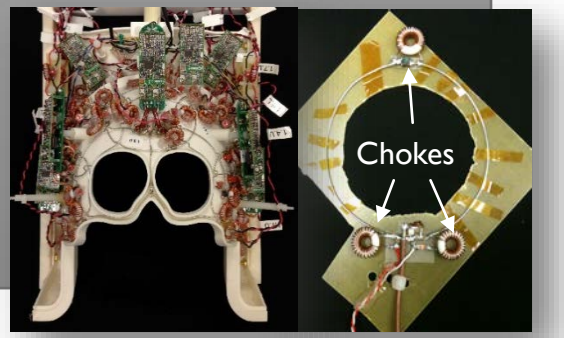
Here: Exploratory work to choose optimized loop design

MOTIVATION

Challenges:

Larger number of capacitors required for 7T → larger number of toroidal bridging chokes

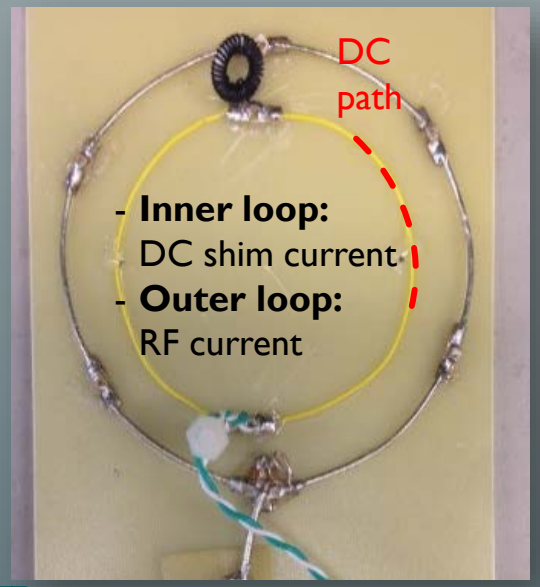
- Additional loss
- Heating
- RF field perturbation
- Construction complexity in confined space



Novel combined conductor concepts:

Coaxial loop element

Concentric loop element

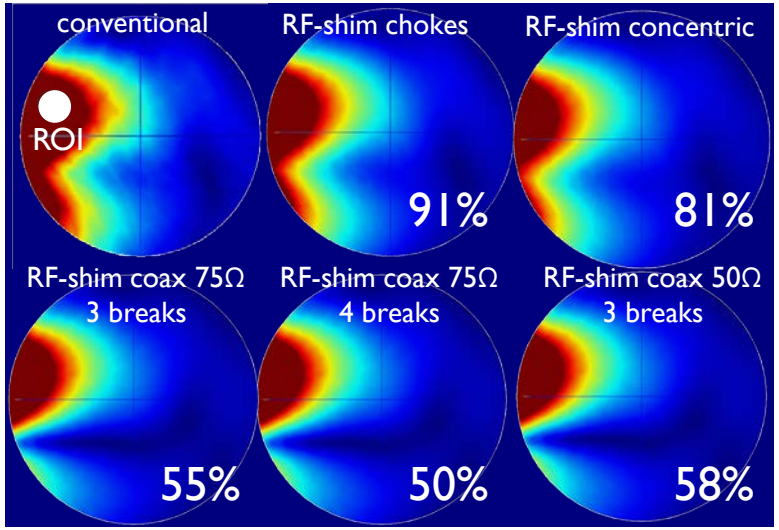


Magnetic coupling between conductors lowers self-inductance of RF loop

→ Reduced number of capacitors
→ No chokes required

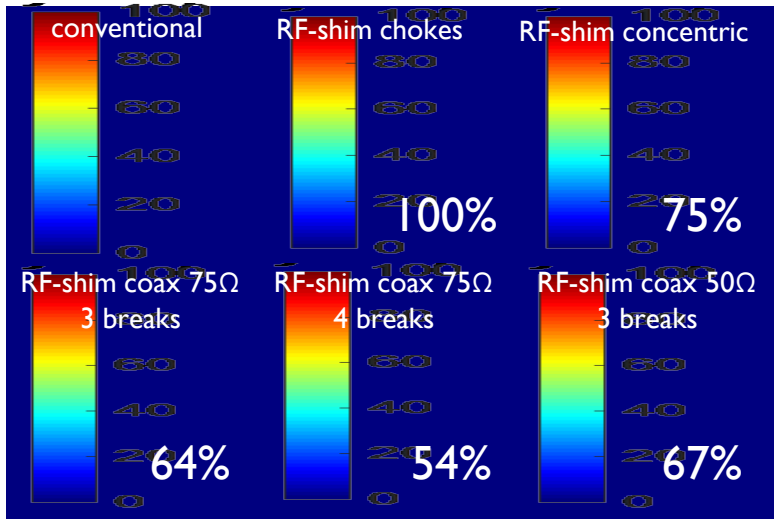
RESULTS - SNR

Simulation

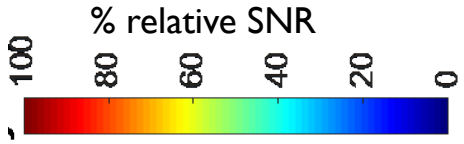


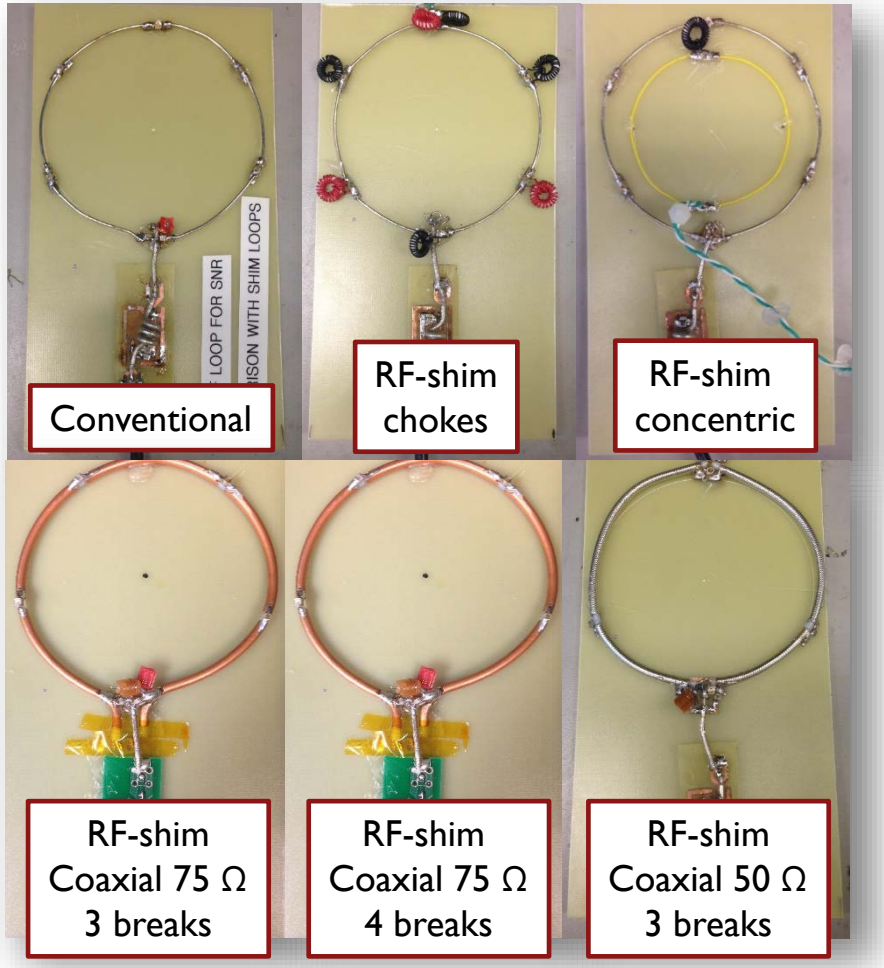
- COMSOL
- Distance to phantom: 1.5 cm

Experiment



- Axial
- TR = 2500 ms
- FOV= 150x150 mm
- 194x194 matrix

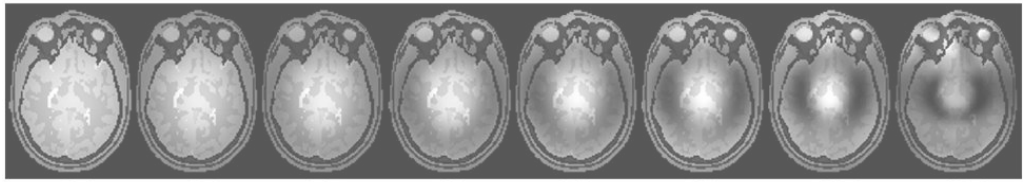




Magna Cum Laude, ISMRM 2015

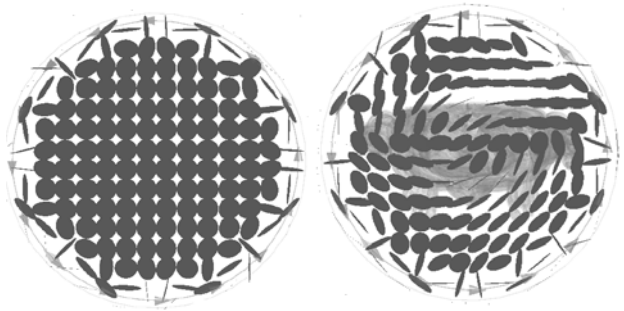
Winkler et al., MRM submitted

HARDWARE RELATED UHF MRI CHALLENGES



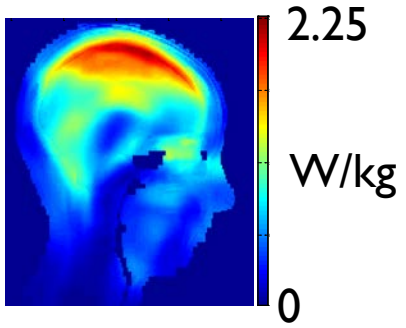
128MHz ····· Larmor frequency ····· 512MHz

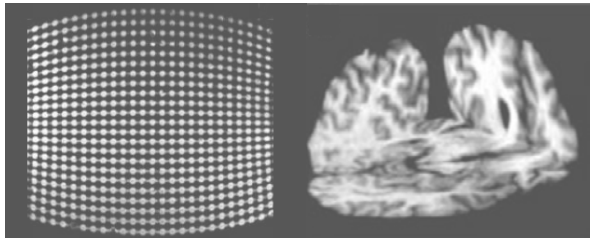
- Global SAR increases with B_0^2
- Local SAR becomes more inhomogeneous due to wave effects
- Local SAR is strongly position- and coil-dependent



B1 inhomogeneities → Image shading

RF heating in tissue (SAR) → Patient safety





B0 inhomogeneities → Geometric distortion & artefacts

High gradient currents → Acoustics/Vibrations & cooling requiments

- Spatial encoding depends on B0 homogeneity
- Signal loss

- Lorentz forces increase with B_0
- Higher sound pressure levels
- Stronger vibrations

- Higher currents require dedicated cooling concepts

MR SAFETY

THERMOACOUSTIC SAR MAPPING – SIMONE WINKLER, PAUL PICOT, MICHAEL THORNTON, BRIAN RUTT

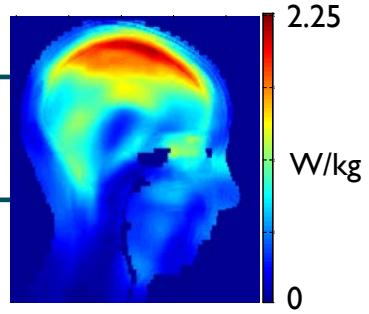
SAR-CONSTRAINED PTX PULSE DESIGN – MIHIR RAJENDRA PENDSE, SIMONE WINKLER, BRIAN RUTT

MR SAFETY BASICS

Challenge:

- Main concern: RF power deposition in tissue (metric: specific absorption rate (SAR))

	Global SAR	Local SAR
	RF power deposition over entire anatomy of interest	local variation of SAR that leads to prominent hotspots
IEC regulation:	3.2 W/kg	10 W/kg



Low field MRI:

Wave effects are not prominent and local SAR is therefore easier to predict

High-field MRI:

1. SAR increases with $B0^2$
2. variation in tissue properties AND fields →strong variations in local SAR – SAR is strongly dependent on:
   - Patient
   - Coil
   - Position
3. pTx techniques may cause worst-case constructive interference of individual local SAR patterns → great concerns!

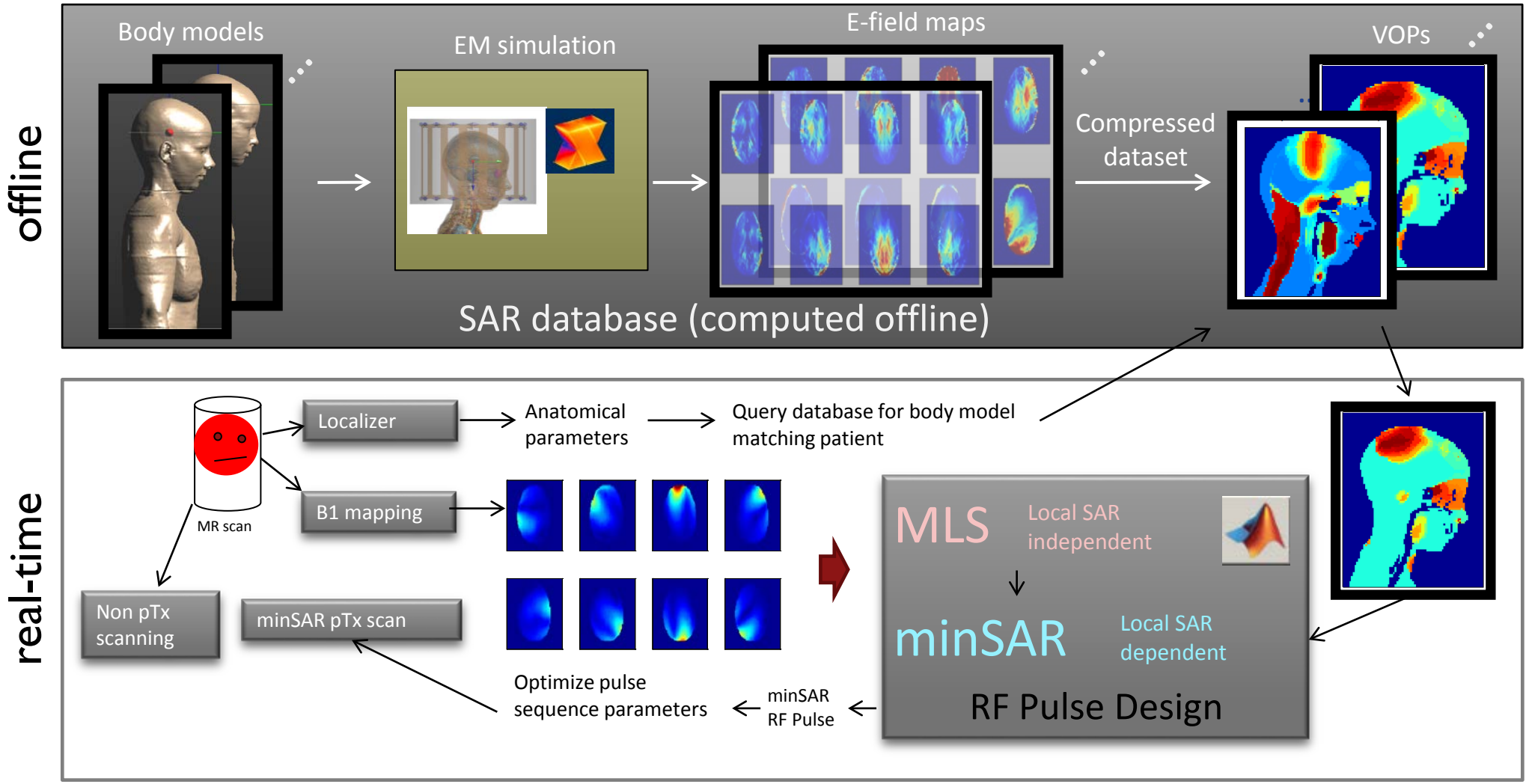
Remedy:

To date there is no experimental procedure to measure local SAR!

- SAR-minimized pTx pulse design
- Find methods to experimentally determine SAR
- Find alternative metric (temperature)

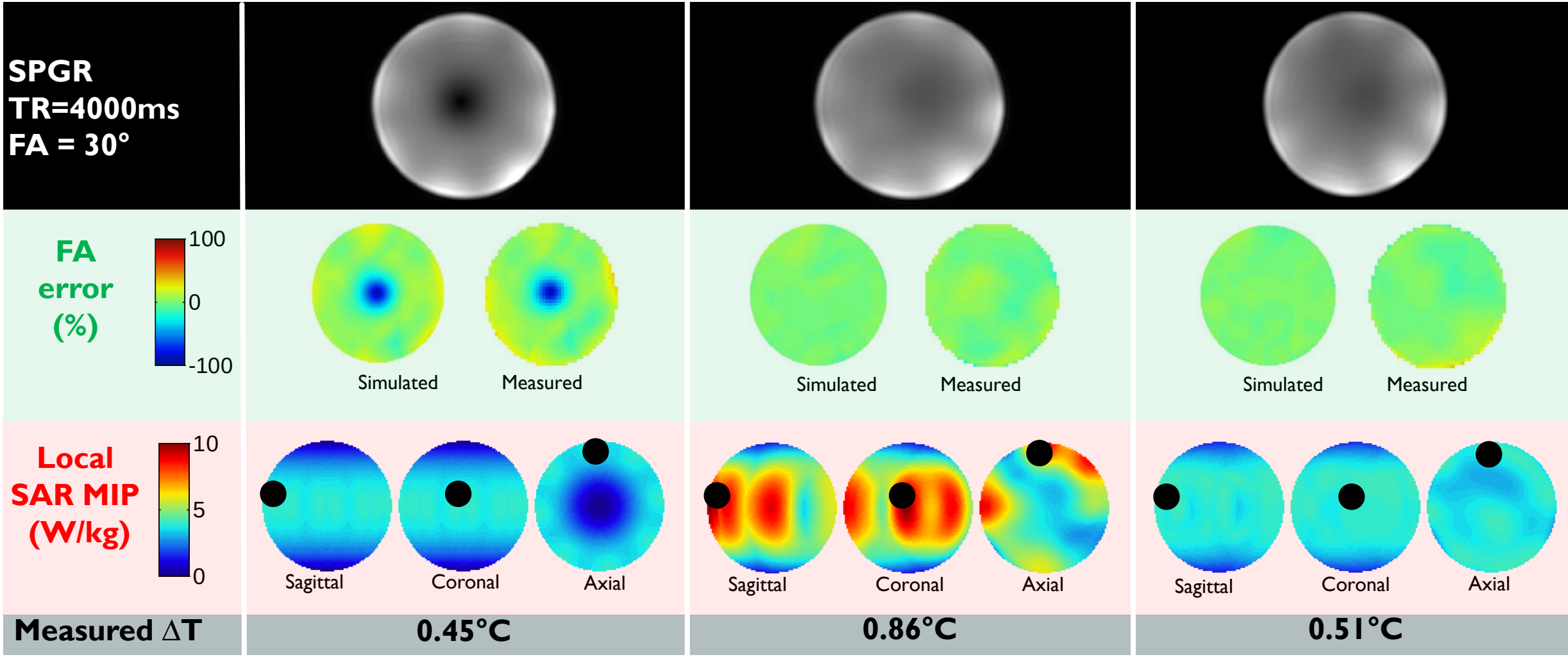
Alternative metric: temperature in tissue (max. 1°C heating)

SAR-MINIMIZED PTX PULSE DESIGN



Pendse, Winkler and Rutt, ISMRM 2014

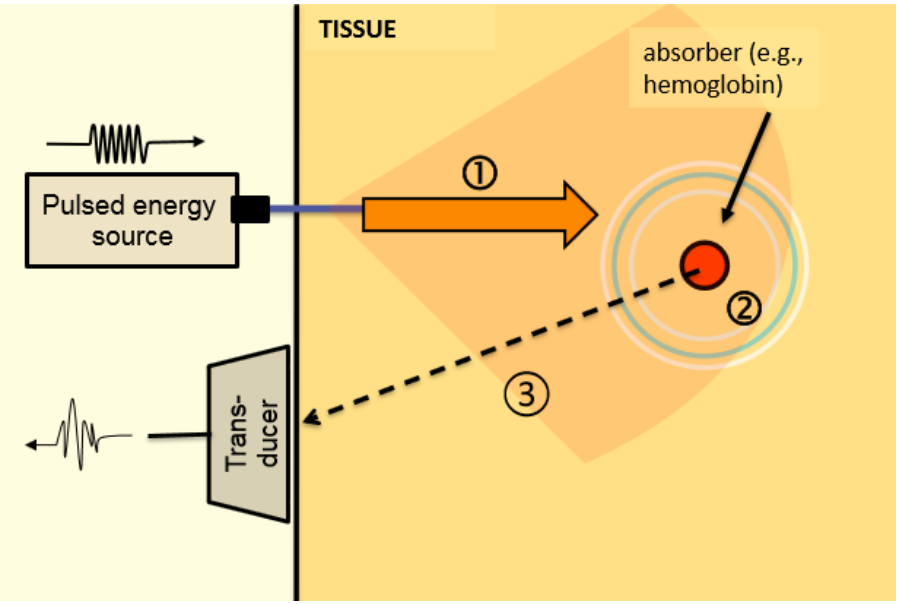
RESULTS: SAR REDUCTION AND FA UNIFORMITY

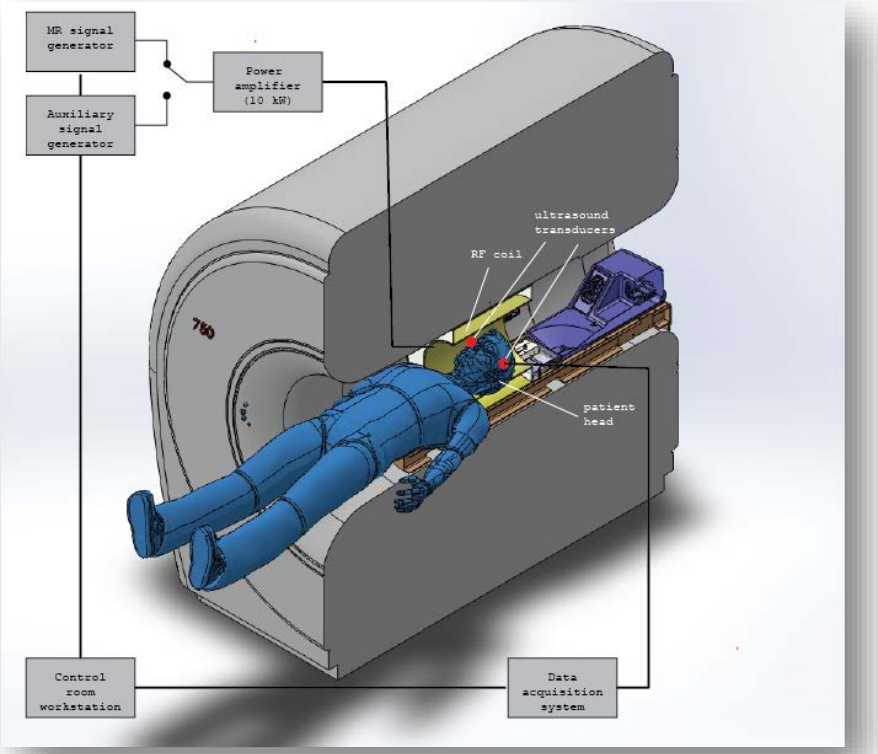


[1]Grissom, MRM 2012;68:1553–1562

Pendse et al., ISMRM 2015 Abstract #543

THERMOACOUSTIC SAR CONCEPT





IDEA:

- absorbed RF energy ≡ SAR!
- MR RF coil as RF transmitter → generates same local SAR pattern as in an MR scan
- RF-induced thermoacoustic signal is proportional to local SAR → use MR coil with short bursts of RF energy

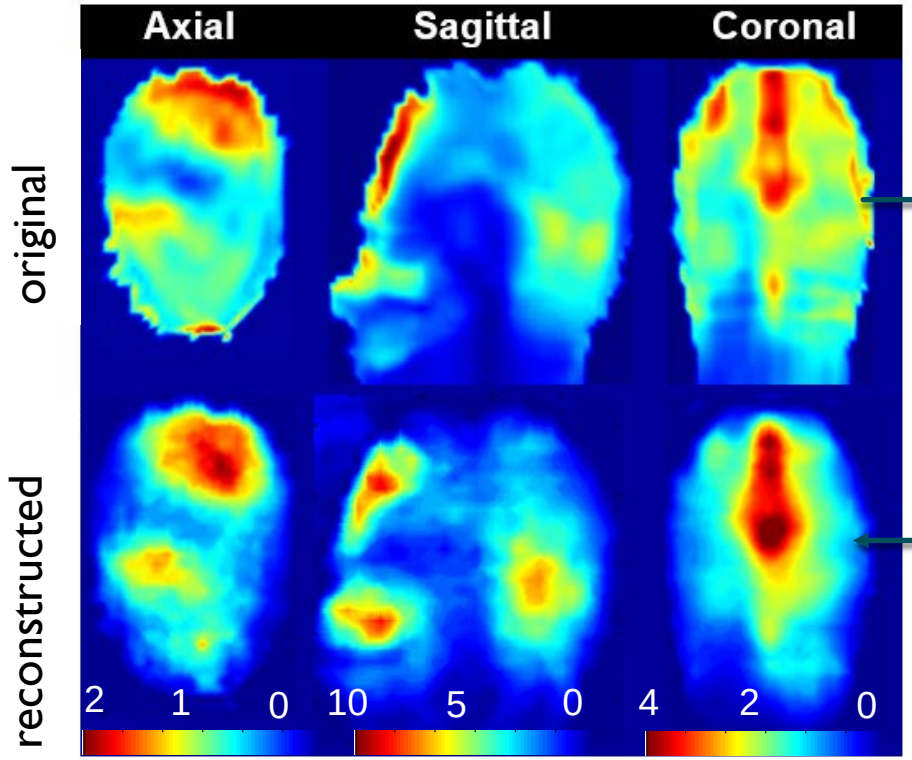
$$\left[\nabla^2 - \frac{1}{v_s{}^2}\frac{\partial^2}{\partial t^2}\right]\boldsymbol{p} = -\frac{\beta\rho}{C_p}\widetilde{\mathbf{SAR}}(\mathbf{r})\frac{\partial I}{\partial t}$$

Acoustic pressure wave — Pulsed RF absorption

Winkler et al, US patent, 2015

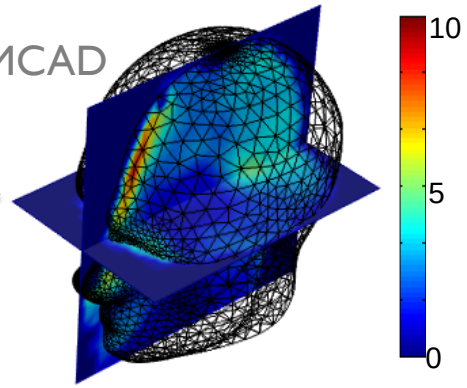
1st place MR engineering, ISMRM 2015

MODELING



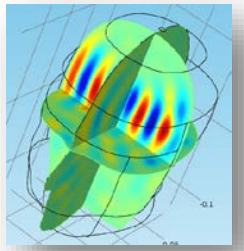
2D SAR pattern in W/kg

SEMCAD



IEC/CENELEC SAM head model with registered SAR pattern from Ella.

COMSOL



$$\left[\nabla^2 - \frac{1}{{v_s}^2}\frac{\partial^2}{\partial t^2}\right] p = -\frac{\beta\rho}{C_p}\widetilde{\mathrm{SAR}}(\mathbf{r})\frac{\partial I}{\partial t}$$

Acoustic pressure wave

Pulsed RF absorption

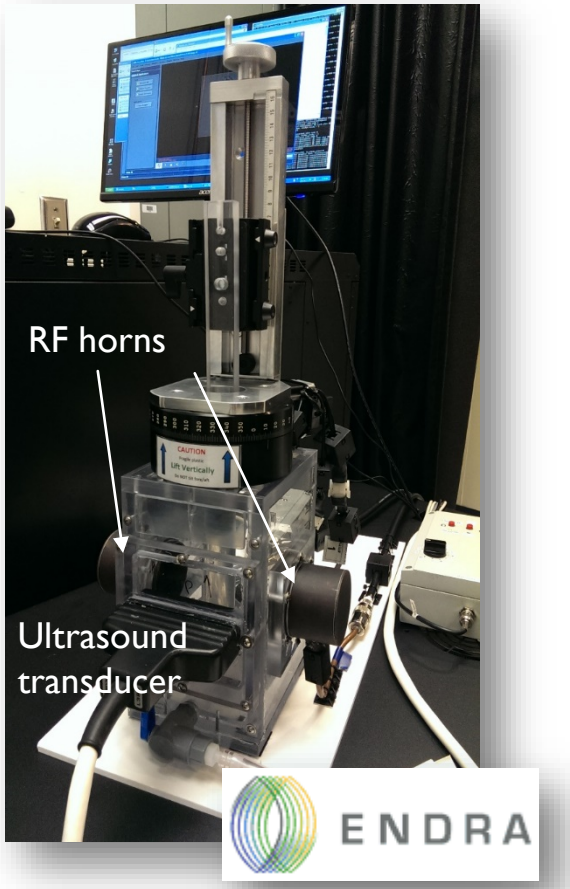
Matlab

Delay-and-Sum reconstruction

Winkler et al., ISMRM 2014, MRM 2016

TEST PLATFORM

- Proof of concept with thermoacoustic imaging platform at Stanford (collaboration with Endra, Inc.)



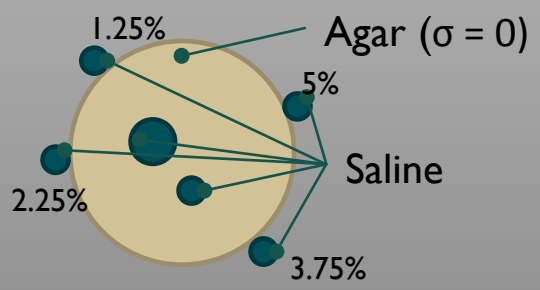
$$\mathrm{SAR} = \int_{\mathrm{sample}} \frac{\sigma(\mathbf{r})|\mathbf{E}(\mathbf{r})|^2}{\rho(\mathbf{r})} d\mathbf{r}$$

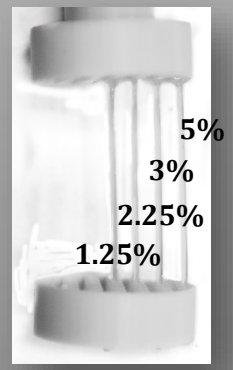
By varying
1. Conductivity
2. Electric Field

in a phantom, we can emulate the local SAR variation in tissue in UHF MRI

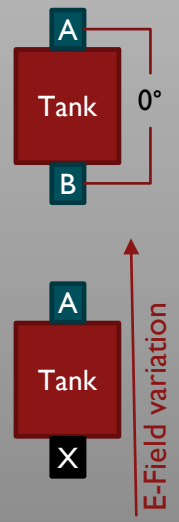
1 – conductivity variation

Point source experiments





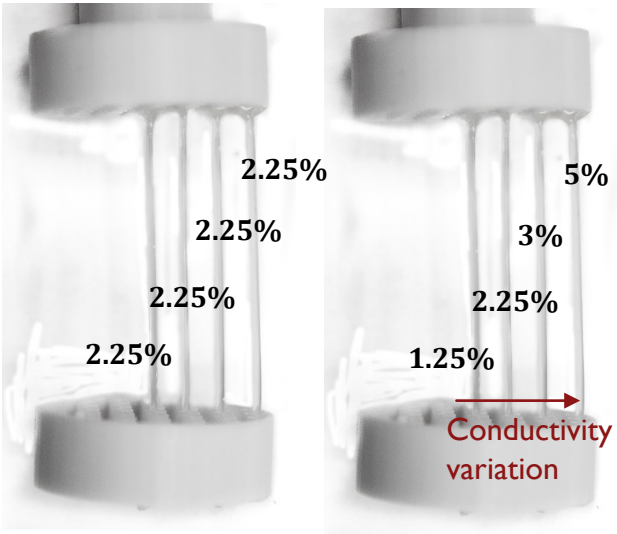
2 – *E*-field variation

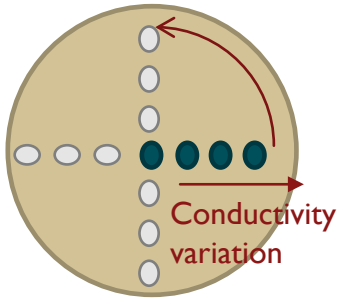


EXPERIMENTS

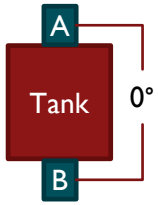
Conductivity variation

Constant conductivity | Varied conductivity

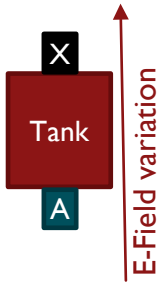




E-Field variation



+ rotated ultrasound transducer to average over perceived *E*-field



Without rotation to maximize *E*-field variation

E-field was measured with a coaxial probe

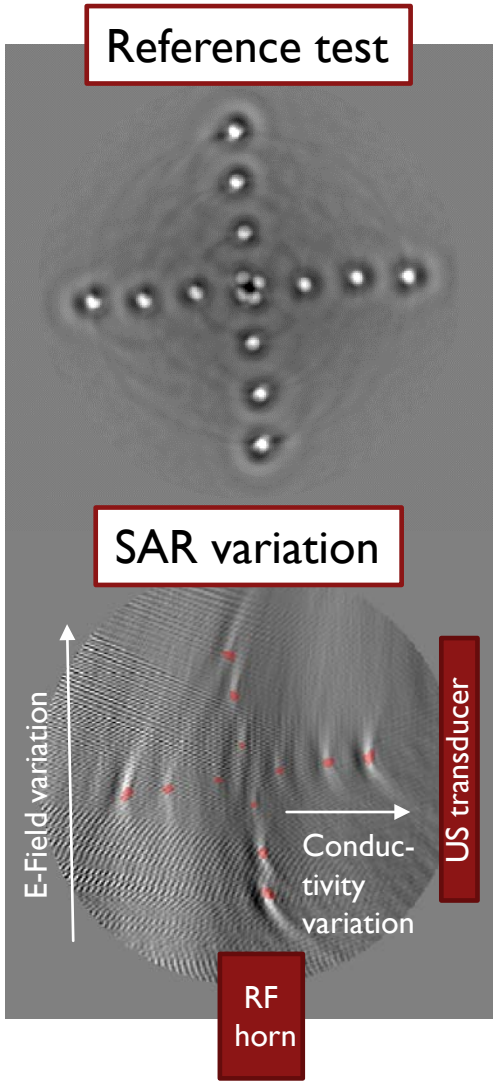
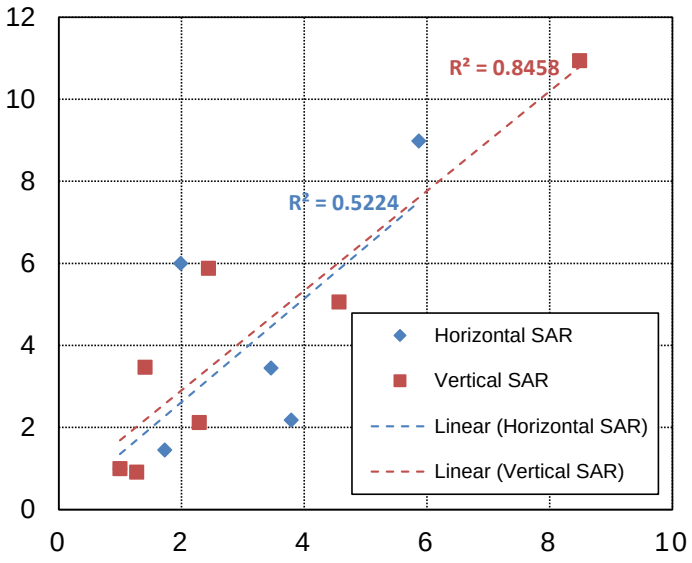


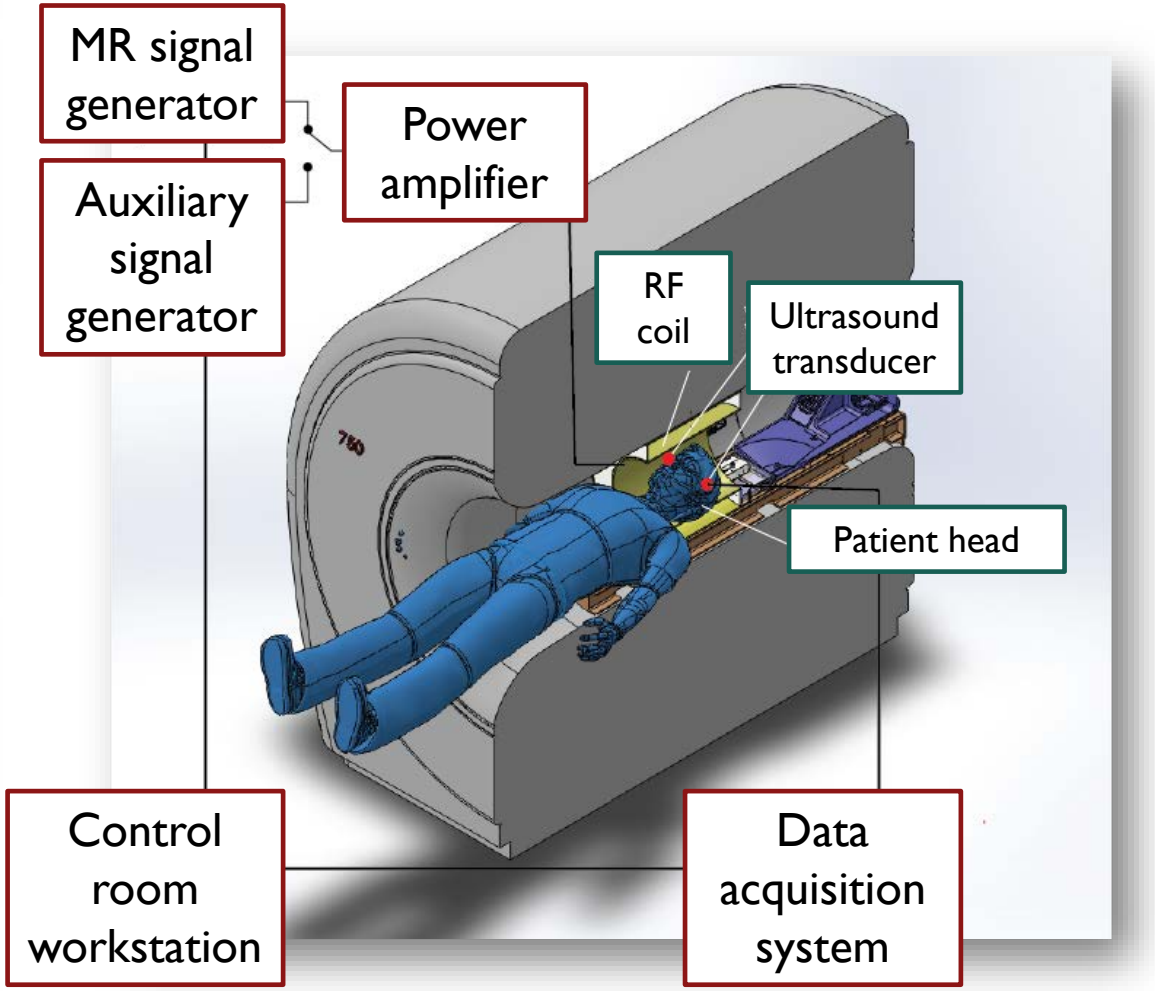
Image values fitted to SAR equation:

$$\mathrm{SAR} = \int_{\mathrm{sample}} \frac{\sigma(\mathbf{r})|\mathbf{E}(\mathbf{r})|^2}{\rho(\mathbf{r})} d\mathbf{r}$$



Winkler et al., ISMRM 2015, MRM 2016

MR SYSTEM INTEGRATION

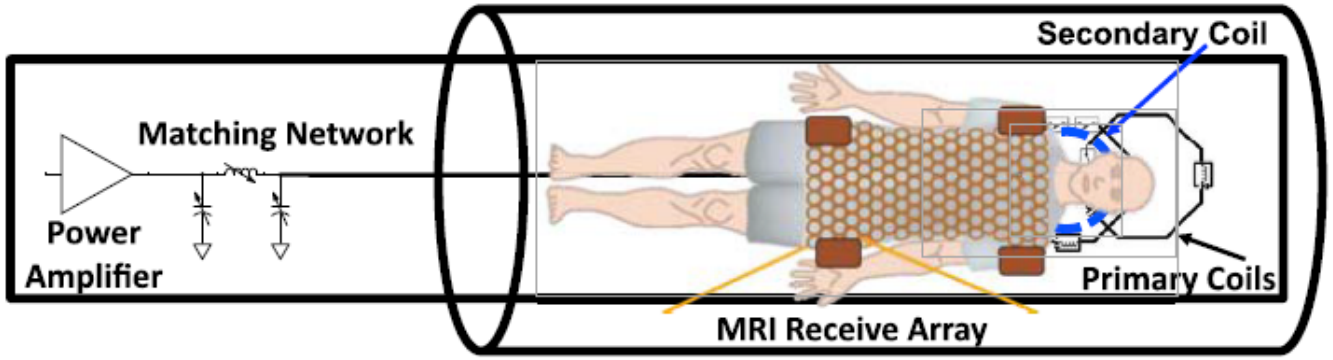


- Fast rise-time pulse shape required → MR system might have to be upgraded with a larger bandwidth auxiliary signal generator
- It is estimated that the power amplifier of the MR system will be delivering sufficient peak power for the experiment
- "Easy" anatomies: SAR in abdomen, knee (7T whole body coil!)
- Brain:
  - An ultrasound bandwidth of 200 kHz will be used for penetrating skull bone
  - Ultrasound transducers will be mounted at two lateral acoustic windows in the skull

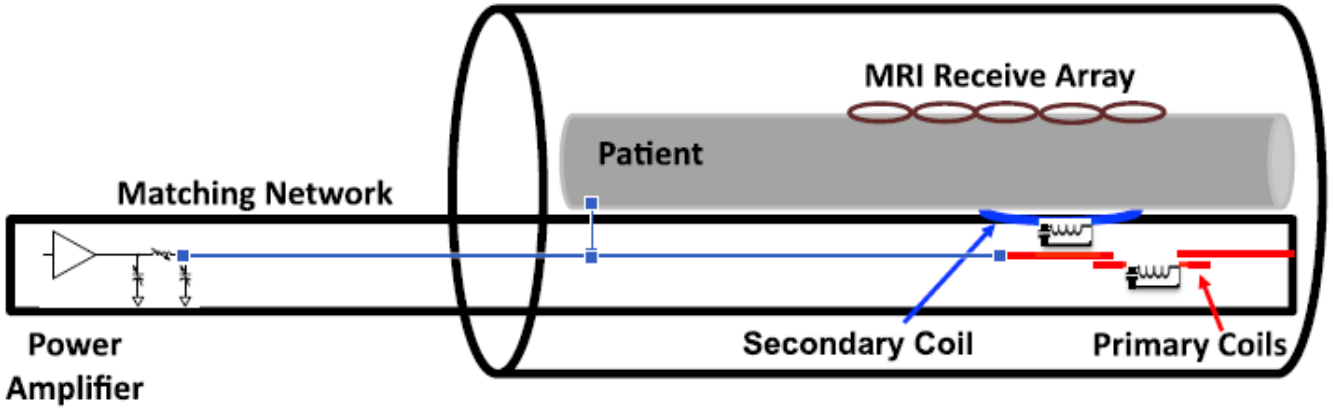
Winkler et al., US Patent, 2015

WIRELESS MRI: POWER TRANSFER

Courtesy Greig Scott, Kelly Byron, Stanford EE



(a)



(b)

Byron et al., IEEE TMTT, 2019

Power transfer from primary to secondary coil to supply receive array

While built at 1.5T, this could be especially useful at UHF because of problems with coil cables

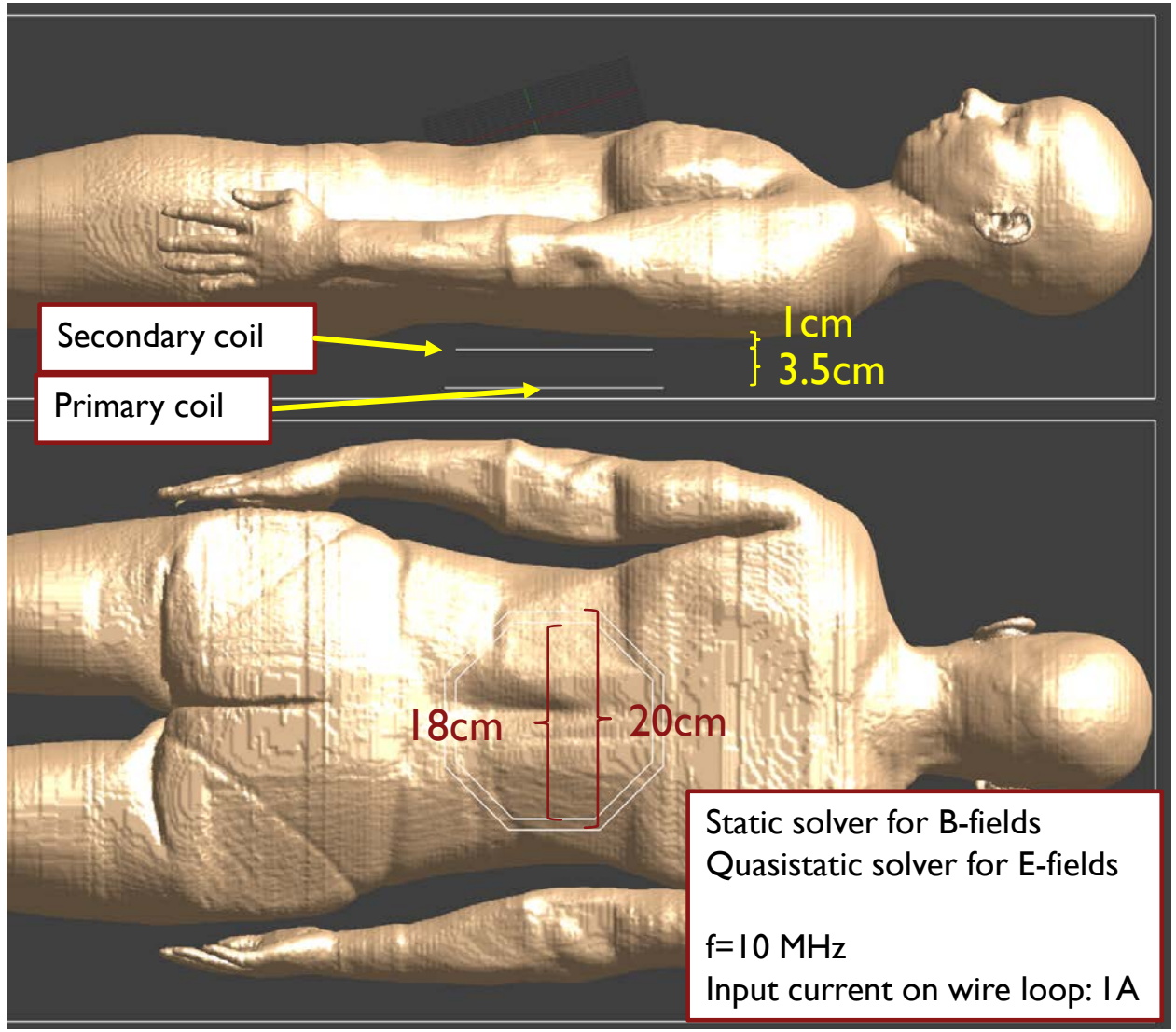
100–300 mW per MRI receive channel

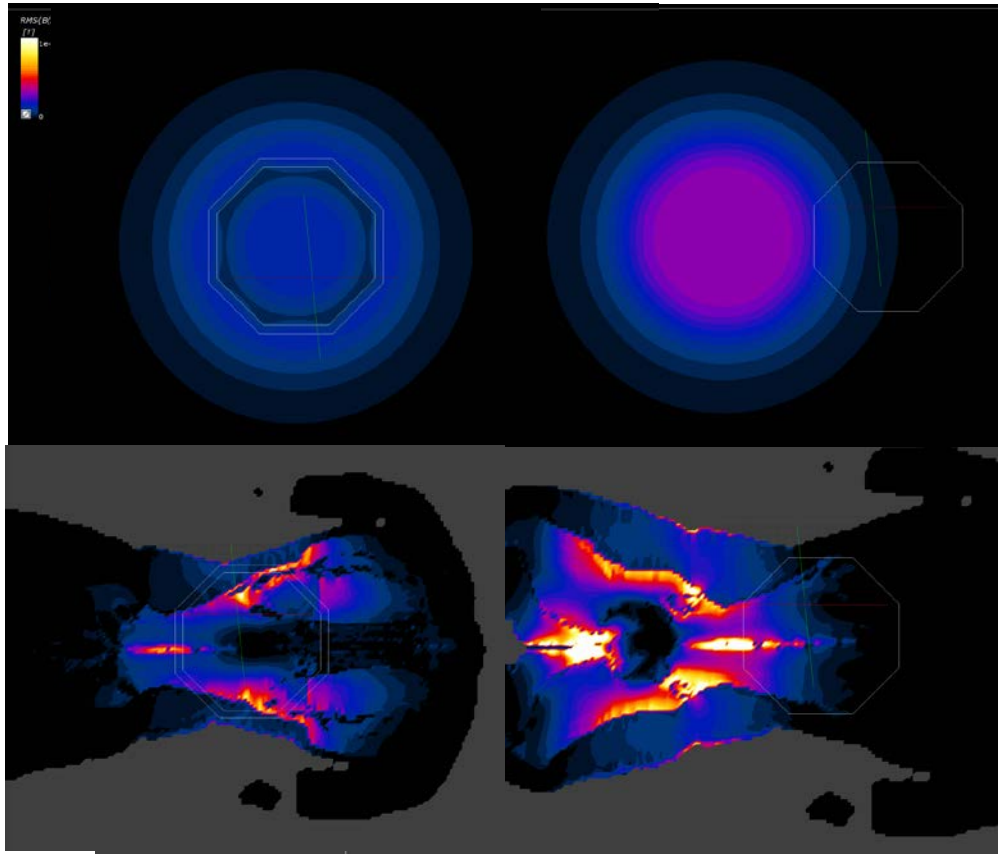
=

3-10 W total

Is this safe for the patient?

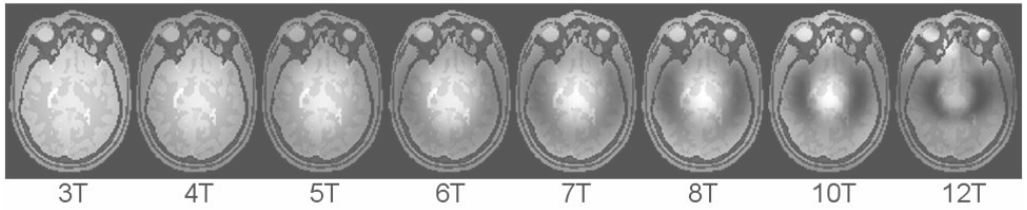
COIL "ILLUMINATION"





Position	Mass averaged SAR	Peak spatial average SAR (1g)	Peak spatial average SAR (10g)
	W/kg	W/kg	W/kg
centered	1.82E-04	7.03E-03	4.41E-03
adjacent	3.48E-04	7.39E-02	1.72E-02

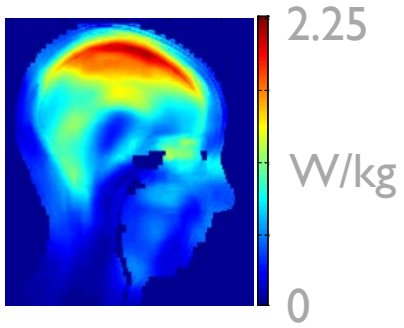
HARDWARE RELATED UHF MRI CHALLENGES

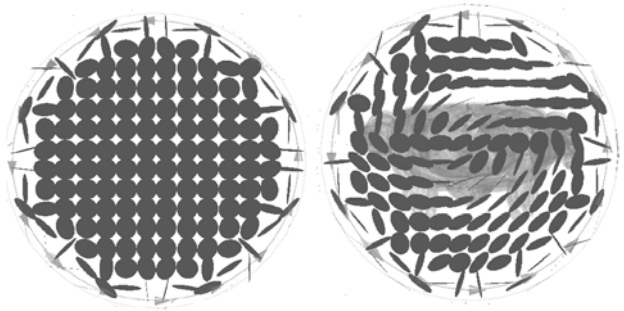


3T 4T 5T 6T 7T 8T 10T 12T

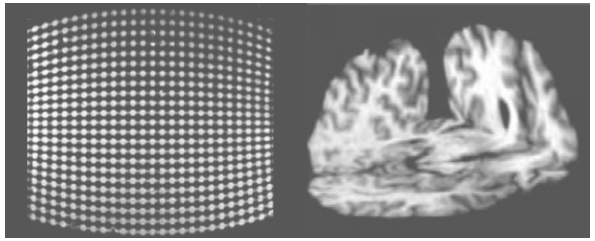
128MHz ····· Larmor frequency ····· 512MHz

- Global SAR increases with B_0^2
- Local SAR becomes more inhomogeneous due to wave effects
- Local SAR is strongly position- and coil-dependent





B1 inhomogeneities → Image shading	RF heating in tissue (SAR) → Patient safety
B0 inhomogeneities → Geometric distortion & artefacts	High gradient currents → Acoustics/Vibrations & cooling requiments



- Spatial encoding depends on B0 homogeneity
- Signal loss

- Lorentz forces increase with B_0
- Higher sound pressure levels
- Stronger vibrations

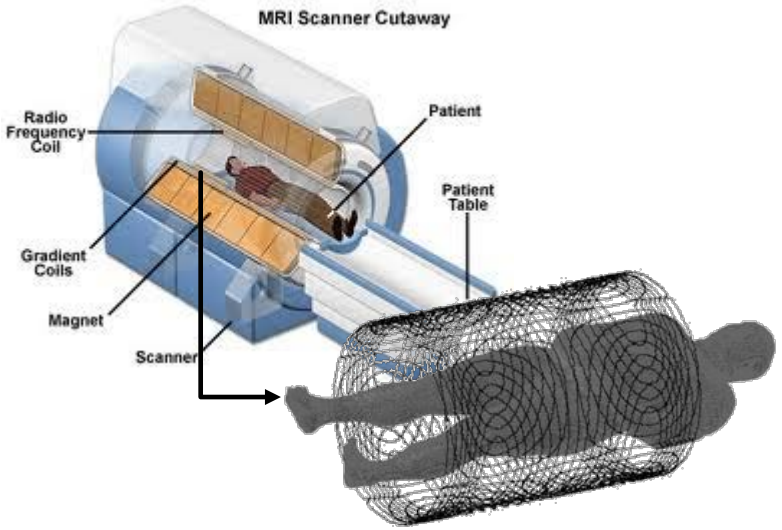
- Higher currents require dedicated cooling concepts

MR ACOUSTICS & THERMAL CONSIDERATIONS

UHF-MRI ACOUSTICS IN HEAD AND BODY GRADIENTS – SIMONE WINKLER, TREVOR WADE, ANDREW ALEJSKI, CHARLES MCKENZIE, BRIAN RUTT

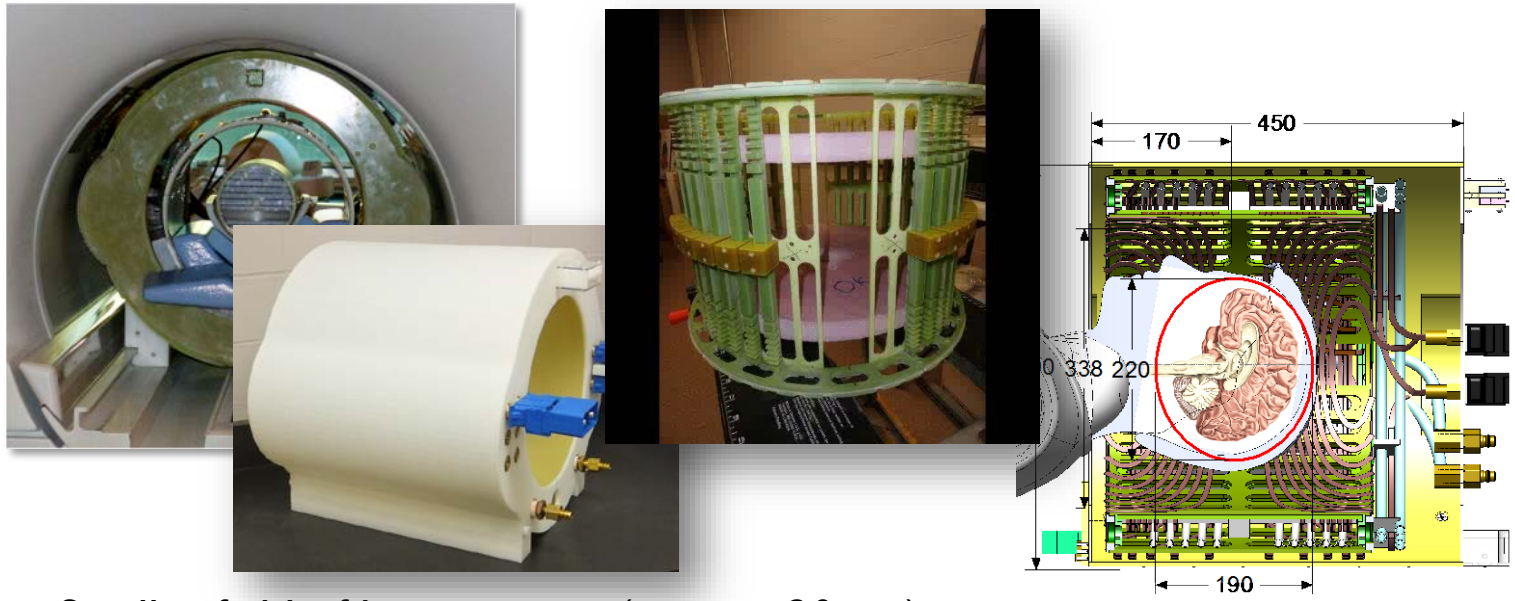
HEAD GRADIENT INSERT

Body gradient coil



- Large linear region (60 cm)
- Low slew rate
- Low gradient strength (50 – 80 mT/m)
- Built into MR system

Head gradient insert coil



- Smaller field of linear region (approx. 20 cm)
- Higher slew rate
- Higher gradient strength (100 – 300 mT/m)
- Inserted into bore

Faster or higher resolution imaging

- Attractive at high field
- **Louder acoustics**

MODELING

Insert gradient modeling:

1. Structural vibration of coil
2. Acoustic wave propagation in air

No displacement in vertical direction

Surrounding air modeling:

Perfectly Matched Layer

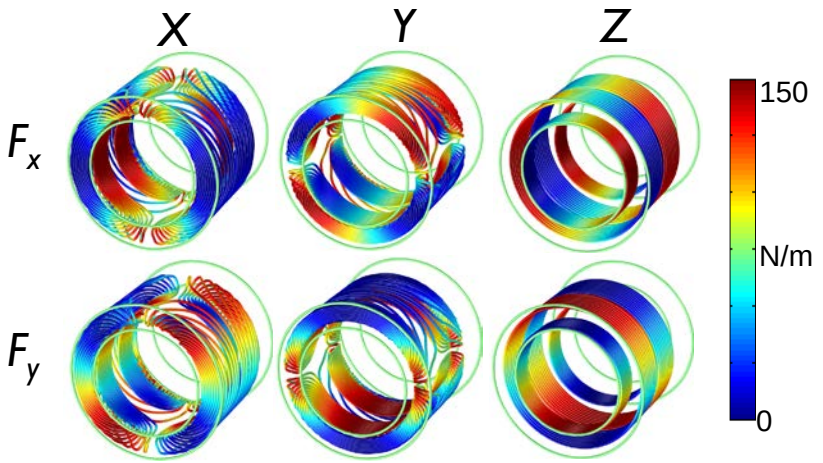
Infinite half space of air on both bore ends

New features in gradient vibroacoustic modeling:
as compared to existing published modalities

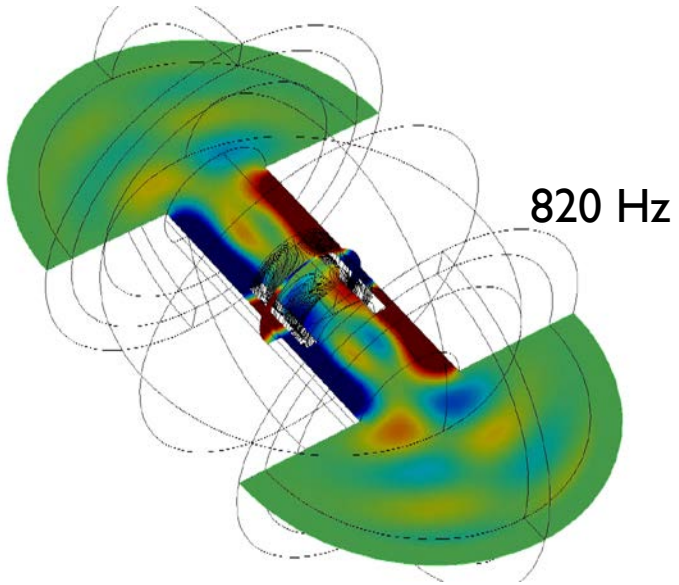
---

1. Accurate wire patterns
2. Realistic bore shape with patient bridge
3. Acoustic propagation outside bore
4. Full coupling of vibrations and acoustics
5. **Lorentz damping**

Lorentz Forces:



Acoustic wave propagation:



Highly accurate prediction of SPL in gradient coils
→ acoustic-based design strategies for SPL reduction become possible

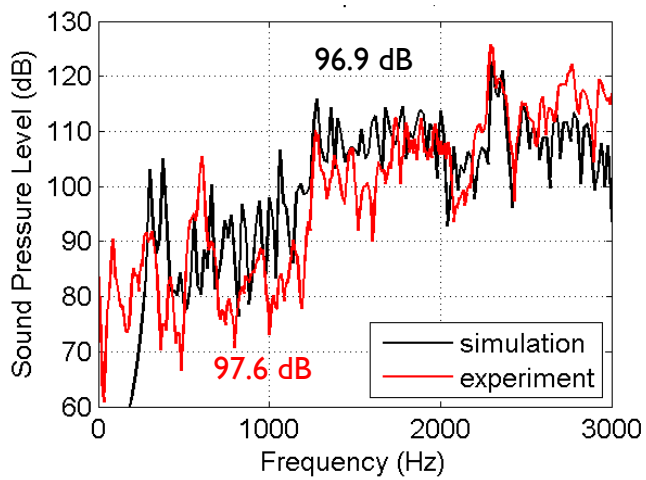
RESULTS REALISTIC HEAD GRADIENT

50A, 3T

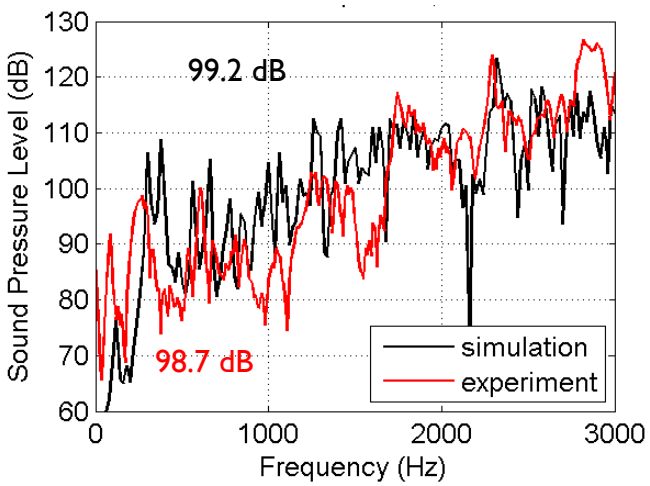
Sound pressure level

isocenter

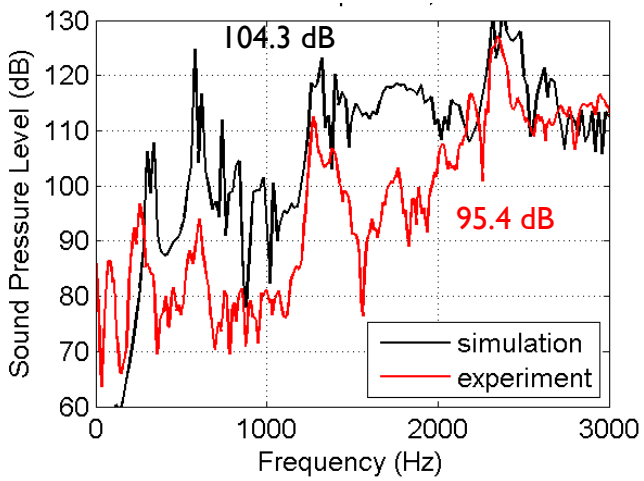
X-gradient



Y-gradient

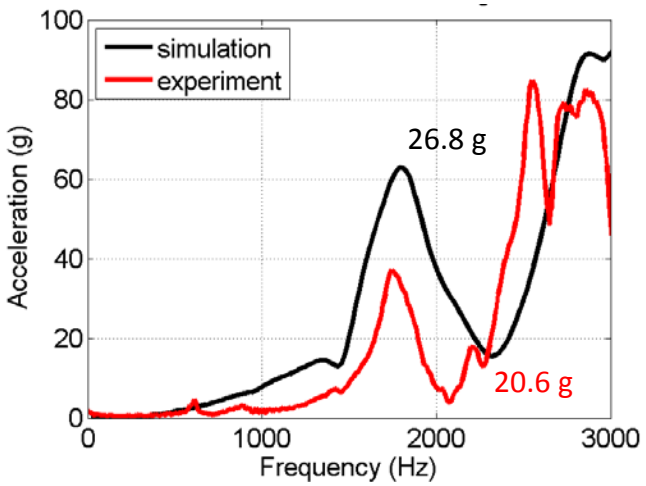


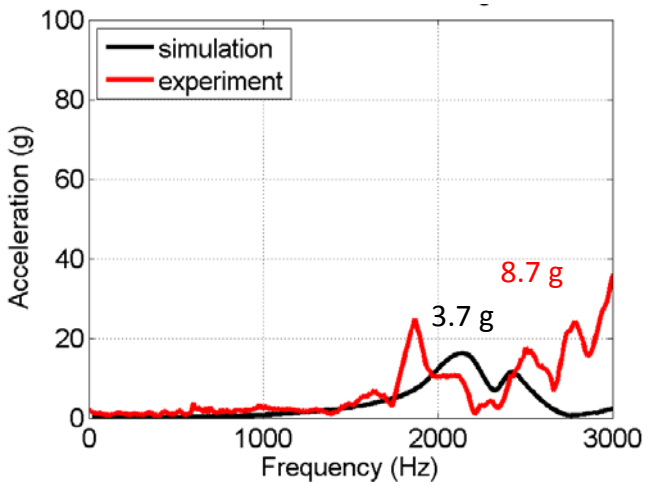
Z-gradient

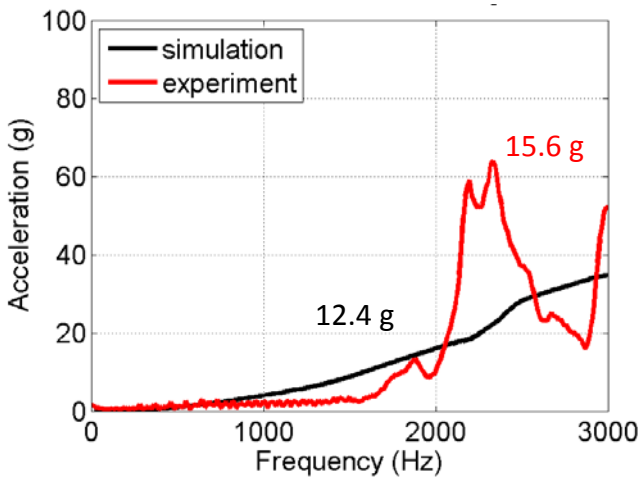


Accelerations

Inner bore wall, +16 cm







Winkler et al., ISMRM 2014

IMPORTANT FINDINGS

Lorentz damping

Eddy current density induced in moving conductor

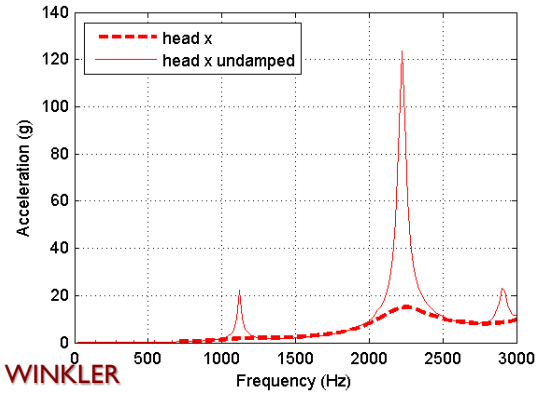
$J_{CL} = \sigma*(v \times B_0)$

Counter-Lorentz force induced due to Eddy current

$F_{CL} = J_{CL}*A \times B_0$

Total force explicit for B_0 in z-direction:

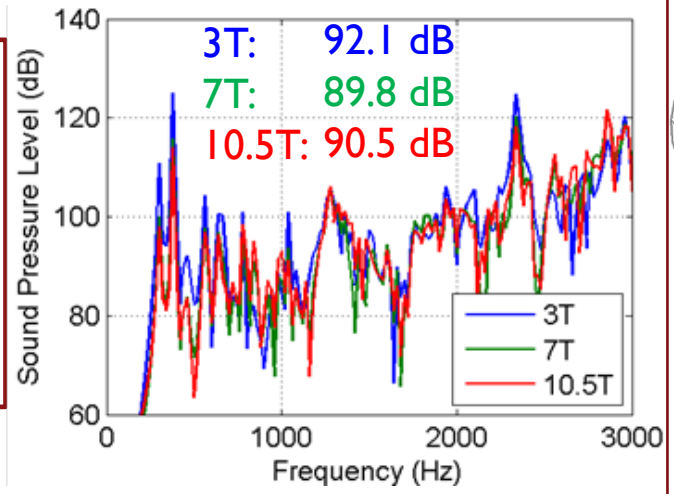
$$F = F_L - \sigma A B_0^2 * v$$



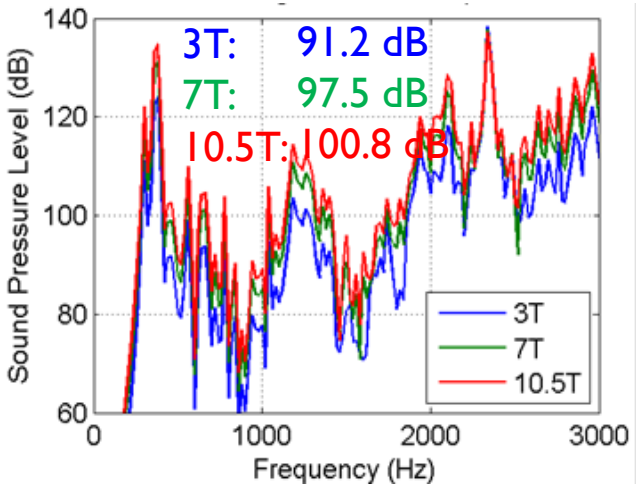
S. A. WINKLER

Field strength dependence

Lorentz damping

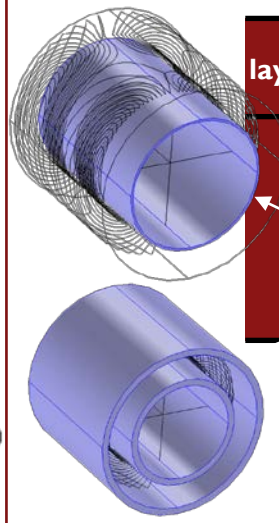


No Lorentz damping

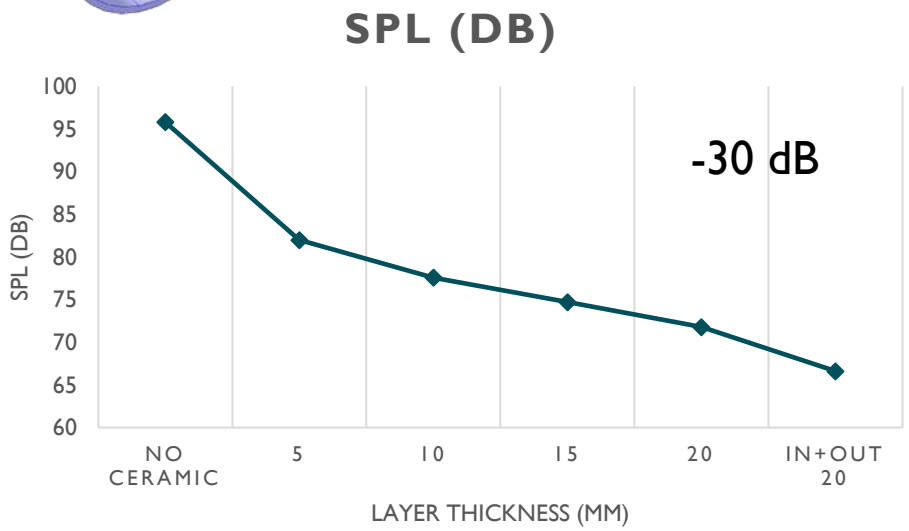


Winkler et al., ISMRM 2015

Ceramic liners

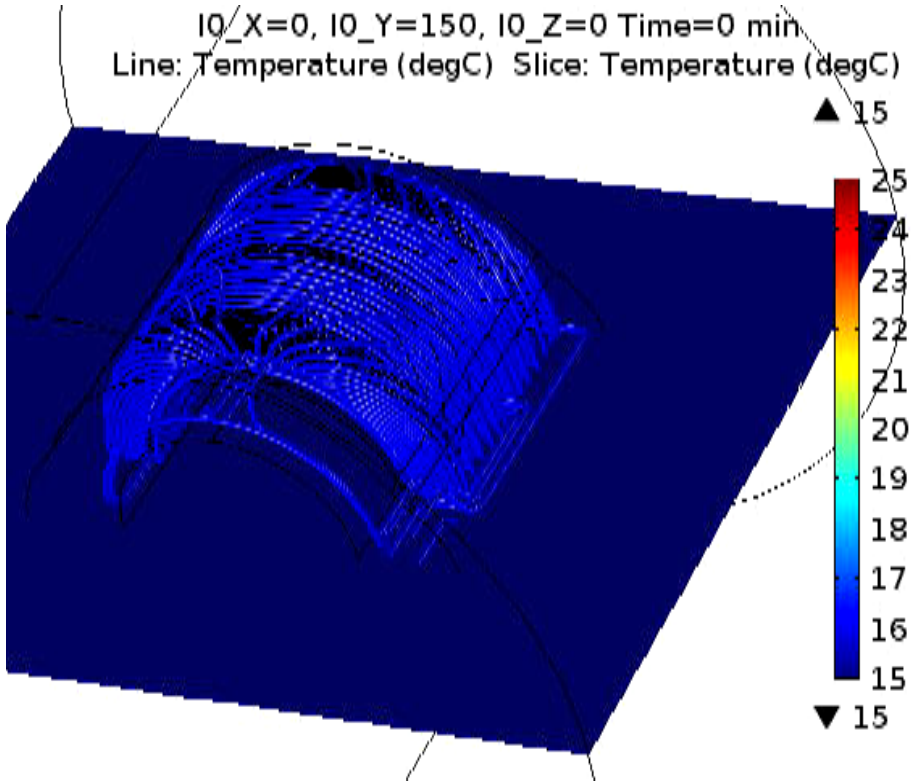


layer thickness (mm)	SPL (dB)	Accel (g)
no ceramic	95.82	4.71
5	82.00	1.29
10	77.60	0.84
15	74.73	0.67
20	71.79	0.56
in+out 20	66.62	0.34



EXAMPLE TEMPORAL PROFILE

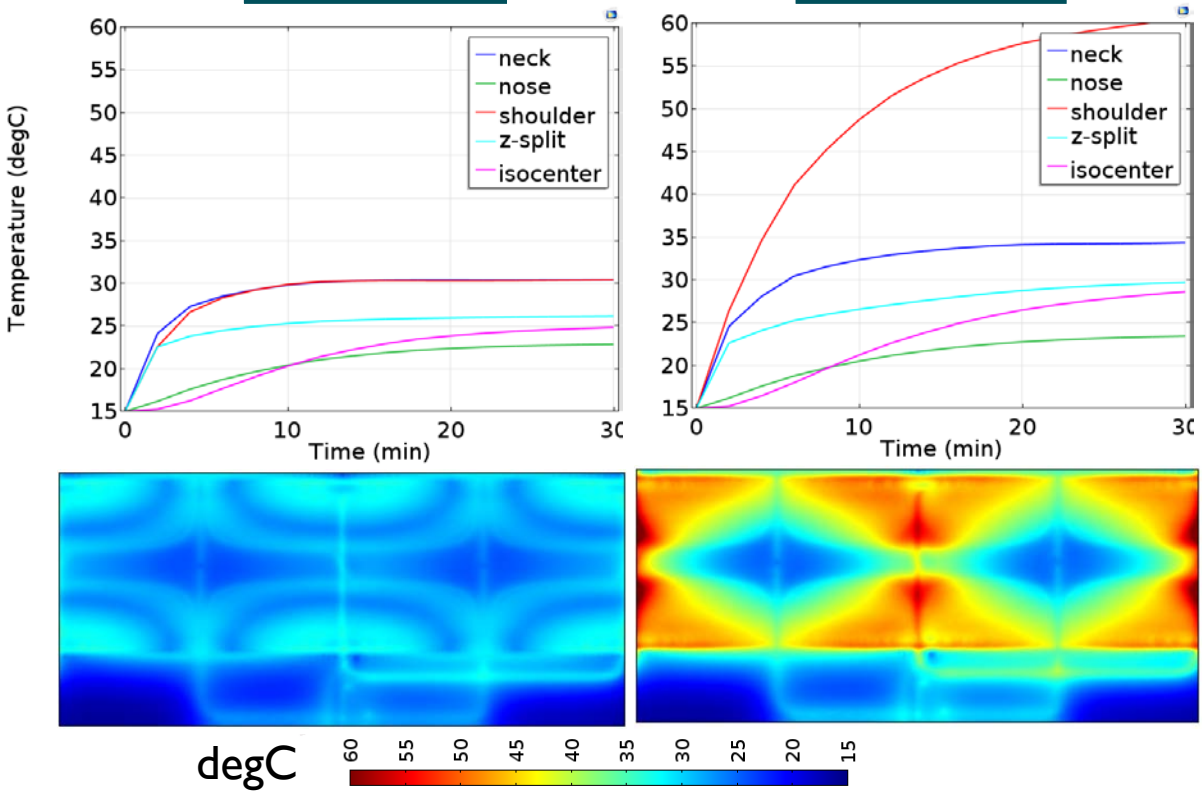
Head gradient insert coil with hollow copper tubing for integrated water cooling



I_{xyz}=150A

Flow x: ON

Flow x: OFF



CONCLUSION

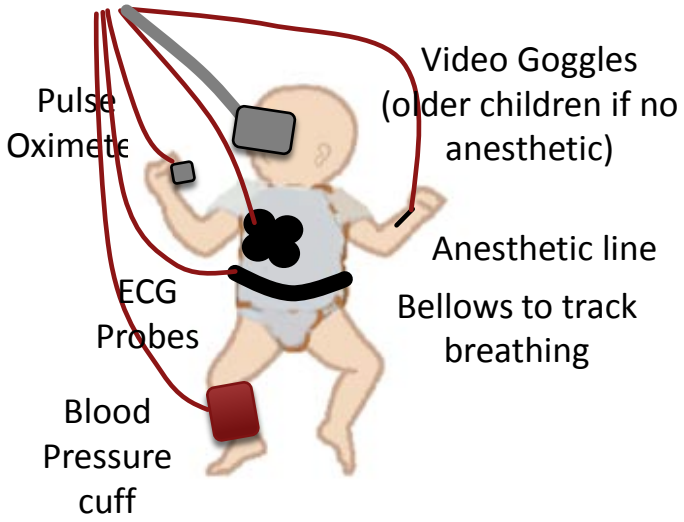
- UHF MRI increases SNR and resolution
- Many technical challenges are faced before clinical adoption
- This talk showed innovative approaches on:
  - B1 inhomogeneities / image shading
  - B0 inhomogeneities
  - MR Safety
  - Acoustics & Thermal considerations

Multiphysics simulations have become an essential tool for UHF MRI engineering!

SIDE TOPIC: FLEXIBLE COILS

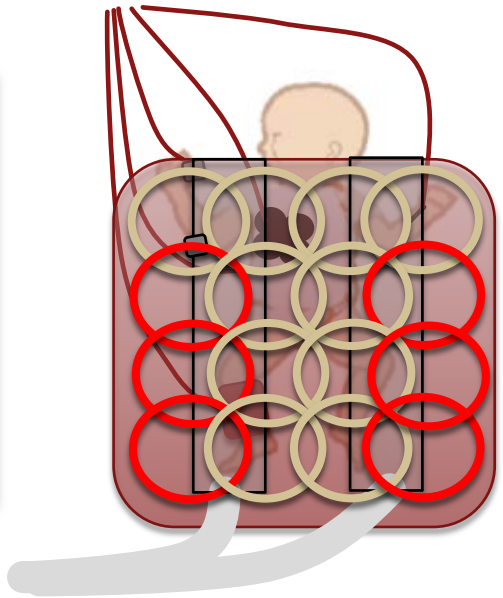
INTRODUCTION / CONCEPT

Pediatric Patient with monitoring equipment

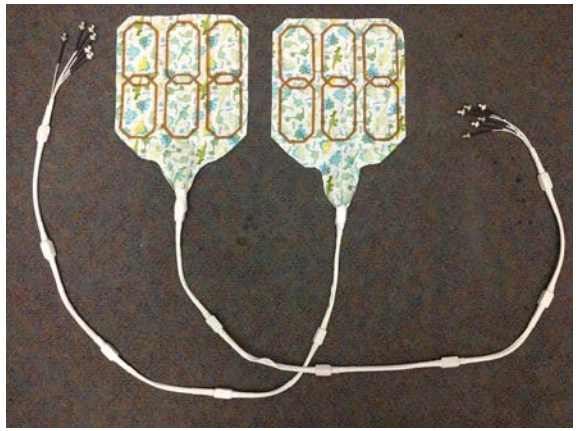


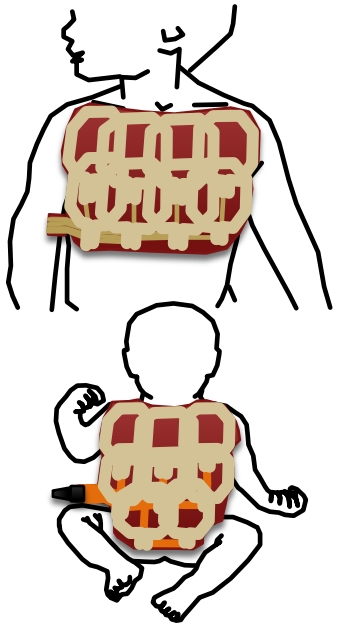
Adult-sized commercial cardiac array:

Large, not all elements contribute useful signal



Solution:
Flexible pediatric-sized screen-printed receive coil array with 12 channels





This work: first clinical pilot study

RESULTS

Caregiver survey

Preference:
1 = commercial adult-sized coil
5 = flexible pediatric coil

Technologists	Anesthesiologists	Nurses	Parents/Child
4.20	3.56	4.07	4.50
4.08 ± 0.62			

Diagnostic quality

Binary scale:
0 = non-diagnostic
1 = diagnostic

100% of cases were diagnostic (1)

Image SNR

Evaluated in lumbar paraspinal muscle

Flexible pediatric array: 53.8±49.3

Commercial adult-sized array: 43.8±13.6

More details on SNR measurements in the E-Poster

Collaborators:

- **Cornell:**
  - Ajay Gupta
  - Kelly Gillen
- **Harvard/MGH:**
  - Larry Wald
  - Jason Stockmann
  - Boris Keil
- **GE:**
  - Fraser Robb
- **Stanford:**
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  - Riccardo Stara
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  - Shreyas Vasanawala
  - John Pauly
  - Kendall O'Brien
  - Kelly Byron
  - Jonathan Lu
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