HOW ULTRA HIGH-FIELD MRI IS REVOLUTIONIZING MEDICINE
Open Engineering and Multiphysics Challenges

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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 $B_0$
2. Perturbation of alignment by use of an RF (electromagnetic) field → magnetization of protons can be measured.
MR BASICS – STEP 1: SPIN ALIGNMENT

**Alignment**

Without external B0-field

With external B0-field

No net magnetization

**Precession**

Spin precession at Larmor frequency

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

\[ \omega_0 = \gamma B_0 \]

\( \gamma \) = Larmor frequency

\( B_0 \) = Static magnetic field strength

Courtesy B. Hargreaves
RF excitation: Additive B1 pulse in transverse plane at Larmor frequency.

Idea: measure magnetic field that spin generates!

Precession at Larmor Frequency

After RF pulse is turned off:

Tissue contrast:

White matter:
- $T1 = 832\text{ms}$
- $T2 = 110\text{ms}$

Gray matter:
- $T1 = 1331\text{ms}$
- $T2 = 80\text{ms}$

3T white matter: $T1=832\text{ms}, T2=110\text{ms}$

3T gray matter: $T1=1331\text{ms}, T2=80\text{ms}$
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**

$v(x) = v_0 + \gamma G_x X / h$

With **linear** additive gradient:

Spatial information can be recovered from Inverse Fourier Transform!!

Gradient coils

DC current generates linear B0 variation

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**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?"

- 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)

Energy gap:

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

NIH K99/R00 funding
Transition to Independence
2017-2022

HCP at the frontier of brain understanding

High-resolution UHF MRI for HCP

Impact

- 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

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

- Spatial encoding depends on $B_0$ homogeneity
- Signal loss

- Lorentz forces increase with $B_0$
- Higher sound pressure levels
- Stronger vibrations
- Higher currents require dedicated cooling concepts
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- RF heating in tissue (SAR)
- Patient safety

- B1 inhomogeneities
- Image shading

- B0 inhomogeneities
- Geometric distortion & artefacts

- High gradient currents
- Acoustics/Vibrations & cooling requirements

2.25 W/kg

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**B1 INHOMOGENEITY BASICS**

**Challenge:**

- 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

**Remedy:**

- **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

2. Modular configuration

3. 3D segmentation

Results

Inter-element coupling

Z-segmentation

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

Stara et al., ISMRM 2014
OPTIMIZED DIELECTRIC SHIMMING

- Electromagnetic scattering is **ANALYTICALLY** described for dielectric-absorptive spheres!
- \( \rightarrow \) 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 vector spherical harmonics

\[
H = \frac{1}{z_0} \sum_{l=0}^{\infty} \sum_{m=-l}^{l} [p(l,m)TM_{lm} + q(l,m)TE_{lm}]
\]
\[
E = \sum_{l=0}^{\infty} \sum_{m=-l}^{l} [q(l,m)TM_{lm} + p(l,m)TE_{lm}],
\]

**Applications:**
- 3T breast MR
- 7T neuroimaging

Winkler et al., Workshop Bremen 2014, Book Chapter 2014
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**Image shading**

**Patient safety**

**B1 inhomogeneities**

**RF heating in tissue (SAR)**

**B0 inhomogeneities**

**High gradient currents**

**Geometric distortion & artefacts**

**Acoustics/Vibrations & cooling requirements**

2.25 W/kg

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B0 INHOMOGENEITIES

RF SHIM COIL AT 7T – SIMONE WINKLER, JASON STOCKMANN, ET AL.
B0 INHOMOGENEITY BASICS

- 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.)

Challenge:

Remedy:

B0 shimming:

- Passive
- Iron inserts
- Active
- Spherical harmonics
- Matrix shim
- 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

**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


Truong TK, Neuroscience 103:235-240 (2014)
RF-Shim Helmet - Concept:

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)

RF-path only

Conventional RF coil

Chokes to bridge capacitor breaks

Added DC path

Shim-RF coil

Stockmann, MRM early view (2015)
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 requirements (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)
Using the RF-shim concept at 7T appears compelling:

1. Increased in-vivo $B_0$ distortion at 7T $\Rightarrow$ 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

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

Optimized loop design becomes essential
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
- Inner conductor: DC shim current
- Outer conductor (RF shield): RF current

Concentric loop element
- Inner loop: DC shim current
- Outer loop: RF current

Magnetic coupling between conductors lowers self-inductance of RF loop
→ Reduced number of capacitors
→ No chokes required
**RESULTS - SNR**

- **COMSOL**
- **Distance to phantom: 1.5 cm**

### Simulation

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

### Experiment

- **100%**
- **75%**
- **64%**
- **54%**
- **67%**

% relative SNR

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**Magna Cum Laude, ISMRM 2015**

Winkler et al., MRM submitted
HARDWARE RELATED UHF MRI CHALLENGES

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**High gradient currents**
- Acoustics/Vibrations & cooling requirements

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**UHF MRI Program**

**HARDWARE RELATED UHF MRI CHALLENGES**

- Spatial encoding depends on B0 homogeneity
- Signal loss

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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

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

- **Global SAR**: RF power deposition over entire anatomy of interest
- **Local SAR**: local variation of SAR that leads to prominent hotspots

**IEC regulation:**
- Low field MRI: 3.2 W/kg
- High-field MRI: 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 $B_0^2$
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!

**Challenge:**

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

**Remedy:**

- 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

**Offline Phase**

1. **Body models**
2. **EM simulation**
3. **E-field maps**
4. **SAR database (computed offline)**

**Real-time Phase**

1. **MR scan**
2. **B1 mapping**
3. **Localizer**
4. **Query database for body model matching patient**

**RF Pulse Design**

- **Non pTx scanning**
- **minSAR pTx scan**
- **Optimize pulse sequence parameters**
- **minSAR RF Pulse**

**Local SAR**

- **Local SAR independent**
- **Local SAR dependent**

**MLS**

**SAR database (computed offline)**

**Compressed dataset**

- **VOPs**

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**Pendse, Winkler and Rutt, ISMRM 2014**
RESULTS: SAR REDUCTION AND FA UNIFORMITY

BIRDCAGE MODE

SAR UNAWARE\(^1\)

IMPULSE\(^2\)

<table>
<thead>
<tr>
<th>SPGR TR=4000ms FA = 30°</th>
</tr>
</thead>
</table>

**FA error (%)**

-100 0 100

Simulated Measured Simulated Measured Simulated Measured

**Local SAR MIP (W/kg)**

0 5 10

Sagittal Coronal Axial Sagittal Coronal Axial Sagittal Coronal Axial

**Measured ∆T**

- 0.45°C

- 0.86°C

- 0.51°C

\(^1\)Grissom, MRM 2012;68:1553–1562  
\(^2\)Pendse et al., ISMRM 2015 Abstract #543
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

\[
\nabla^2 - \frac{1}{v_s^2 t^2} \frac{\partial^2}{\partial t^2} p = -\frac{\beta \rho}{C_p} \frac{\partial I}{\partial t} \quad \text{SAR(r)} \frac{\partial I}{\partial t}
\]

Acoustic pressure wave

\[\text{Pulsed RF absorption}\]

Winkler et al, US patent, 2015

S. A. WINKLER

1st place MR engineering, ISMRM 2015
**Modeling**

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

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

Acoustic pressure wave

Pulsed RF absorption

2D SAR pattern in W/kg

Winkler et al., ISMRM 2014, MRM 2016
Proof of concept with thermoacoustic imaging platform at Stanford (collaboration with Endra, Inc.)

\[
SAR = \int_{\text{sample}} \frac{\sigma(r)|E(r)|^2}{\rho(r)} \, dr
\]

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

- Agar (\(\sigma = 0\))
- Saline

2 – E-field variation

- 1) Original
- 2) One excitation channel off
**EXPERIMENTS**

- **Conductivity variation**
  - Constant conductivity
  - Varied conductivity
  - 2.25%
  - 5%
  - 3%

- **E-Field variation**
  - + rotated ultrasound transducer to average over perceived E-field
  - Without rotation to maximize E-field variation

- **Reference test**

- **Image values fitted to SAR equation**:
  \[
  \text{SAR} = \int_{\text{sample}} \frac{\sigma(r)|E(r)|^2}{\rho(r)} \, dr
  \]

- **E-field was measured with a coaxial probe**

- Winkler et al., ISMRM 2015, MRM 2016
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

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?

Byron et al., IEEE TMTT, 2019

S. A. WINKLER

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INTRODUCTION

COIL “ILLUMINATION”

Secondary coil

Primary coil

1 cm

3.5 cm

18 cm

20 cm

Static solver for B-fields
Quasistatic solver for E-fields

f=10 MHz
Input current on wire loop: 1A

SAR (W/kg)

Position

Primary coil

Secondary coil

Static solver for B-fields
Quasistatic solver for E-fields

f=10 MHz
Input current on wire loop: 1A

<table>
<thead>
<tr>
<th>Position</th>
<th>Mass averaged SAR (1g)</th>
<th>Peak spatial average SAR (1g)</th>
<th>Peak spatial average SAR (10g)</th>
</tr>
</thead>
<tbody>
<tr>
<td>centered</td>
<td>1.82E-04</td>
<td>7.03E-03</td>
<td>4.41E-03</td>
</tr>
<tr>
<td>adjacent</td>
<td>3.48E-04</td>
<td>7.39E-02</td>
<td>1.72E-02</td>
</tr>
</tbody>
</table>
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RF heating in tissue (SAR)

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B1 inhomogeneities

Image shading

B0 inhomogeneities

Geometric distortion & artefacts

High gradient currents

Acoustics/Vibrations & cooling requirements

- Lorentz forces increase with $B_0$
- Higher sound pressure levels
- Stronger vibrations
- Higher currents require dedicated cooling concepts

Spatial encoding depends on B0 homogeneity
Signal loss
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

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

- 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:

- Structural vibration of coil
- Acoustic wave propagation in air

Surrounding air modeling:

- Perfectly Matched Layer

---

**New features in gradient vibroacoustic modeling:**

- Accurate wire patterns
- Realistic bore shape with patient bridge
- Acoustic propagation outside bore
- Full coupling of vibrations and acoustics
- **Lorentz damping**

No displacement in vertical direction

Infinite half space of air on both bore ends

---

Highly accurate prediction of SPL in gradient coils

→ acoustic-based design strategies for SPL reduction become possible

---

**Acoustic wave propagation:**

- **820 Hz**

---

**Lorentz Forces:**

- **$F_x$**
- **$F_y$**
RESULTS REALISTIC HEAD GRADIENT

X-gradient

Y-gradient

Z-gradient

Sound pressure level

Isocenter

Frequency (Hz)

Sound Pressure Level (dB)

50A, 3T

Accelerations

Inner bore wall, +10 cm

Frequency (Hz)

Acceleration (g)

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

\[ F_{CL} = J_{CL} \times A \times B_0 \]

**Total force explicit for** \( B_0 \) **in** \( z \)-direction:

\[ F = F_L - \sigma A B_0^2 v \]

**Field strength dependence**

**Ceramic liners**

<table>
<thead>
<tr>
<th>layer thickness (mm)</th>
<th>SPL (dB)</th>
<th>Accel (g)</th>
</tr>
</thead>
<tbody>
<tr>
<td>no ceramic</td>
<td>95.82</td>
<td>4.71</td>
</tr>
<tr>
<td>5</td>
<td>82.00</td>
<td>1.29</td>
</tr>
<tr>
<td>10</td>
<td>77.60</td>
<td>0.84</td>
</tr>
<tr>
<td>15</td>
<td>74.73</td>
<td>0.67</td>
</tr>
<tr>
<td>20</td>
<td>71.79</td>
<td>0.56</td>
</tr>
<tr>
<td>in+out 20</td>
<td>66.62</td>
<td>0.34</td>
</tr>
</tbody>
</table>

\( 3T: 91.2 \text{ dB} \)

\( 7T: 97.5 \text{ dB} \)

\( 10.5T: 100.8 \text{ dB} \)

-30 dB

Winkler et al., ISMRM 2015
Head gradient insert coil with hollow copper tubing for integrated water cooling

Flow x: ON

Flow x: OFF

$T_{xyz}=150\,\text{A}$
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
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

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

<table>
<thead>
<tr>
<th>Caregivers</th>
<th>Preference</th>
<th>Mean ± SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Technologists</td>
<td>4.20</td>
<td>4.08 ± 0.62</td>
</tr>
<tr>
<td>Anesthesiologists</td>
<td>3.56</td>
<td></td>
</tr>
<tr>
<td>Nurses</td>
<td>4.07</td>
<td></td>
</tr>
<tr>
<td>Parents/Child</td>
<td>4.50</td>
<td></td>
</tr>
</tbody>
</table>

Preference: 1 = commercial adult-sized coil, 5 = flexible pediatric coil
THANK YOU!

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