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Magnetic Resonance Imaging

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October 6, 2020

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MRI SYSTEM PRINCIPLE



MRI PRINCIPLE (1/5)

- NMR measures magnetization of atomic nuclei in the presence of magnetic field
- Particles with mass (proton) spin on their axis at Larmor frequency
- Signals are obtained from the NMR observation of proton in water





Larmor frequency ($\omega = \gamma$ Bo) ω is the precession frequency (Hz) γ is the gyromagnetic ratio

Spin precession around Bo direction

No field: random orientation of spins



Proton spins -> microscopic magnetic moment "The human body can be seen as a set of small magnets"



Under field: alignment of spins



MRI PRINCIPLE (2/5)

Double fonctions of RF antenna

Step 1: The antenna send a RF signal (generated by powering a small coil with AC current)

The magnetic moment is depolarized



Step 2: Relaxation to come back to the initial magnetic moment







The relaxation generates a RF signal collected by the antenna





But for a constant background field, signal is identical for every spin! No way to distinguish the different signals...





MRI PRINCIPLE (4/5)

Gradient coils are needed to modify the magnetic field locally





1 Helmoltz coil pairs for the z-direction

2 pairs of saddle coils for the x or y axes



- A given gradient is imposed in one direction during the relaxation
- Precession frequency is now different for each spin
- This enables localisation of image slices as well as phase and frequency encoding





MRI MILESTONES A 80 YEARS STORY...

1937 Rabi (1944) - resonance method for recording magnetic properties of atomic nuclei

1940 Zavoyski – discovery of electron paramagnetic resonance

1946 Block, Purcell (1952) – nuclear magnetic precision measurements, related discoveries 1973 Lauterbur (2003) – First MR images on samples

1977 Mansfield (2003) – First clinical MR images 1978 Philips 0.15T MR scanner 1979 Siemens 0.2T MR scanner

1981 Superconducting MRI scanners (0.5 T, Oxford)

1983 GE generates images with 1.5T scanner

1986 Actively-shielded superconducting scanners

1991 fMRI invented – (15 yrs after first clinical images)

1993 Philips: Compact, actively shielded, no LN shield scanners

1994 Diffusion Tensor Imaging invented

1997 GE introduces ZBO scanners: no LHe refill over lifetime

2000 Commercial 3 T MRI from GE, Siemens and Philips

2001 GE, Philips: High-field Open MRI systems

2005 Siemens: wide-bore cylindrical scanners (70-cm patient bore)

2020 120 million MRI scans per year

Rabi (1944): Nobel prize of 1944



Field uniformity and stability

- Design Uniformity: 10 parts-per-million (ppm) in ~50 cm diameter volume
 - Multiple-coil configuration
 - Sweet spot
- Field decay:
 - short-term decay: 1 ppb during sequence (EMI, vibration)
 - Long-term decay: less than 0.1 ppm/hour on average, less than 0.1% per year

Shielding

 Magnetic field outside of the scanning room shall be less than 5 gauss (industry standard) **DE LA RECHERCHE À L'INDUSTR**



MRI MAGNETIC DESIGN OPTIMIZATION



A complex problem...

- Current transport capacity (choice of conductor)
- Spatial homogeneity
- Fringe field
- Peak field on the conductor
- Cooling mode
- Mechanical stresses
- Manufacturing techniques
- Economical constraints (€)



Inside a sphere with a center O and radius r_{max} « magnetically » empty, the B_z component of the magnetic field can be expressed using a spherical harmonic expansion based on Legendre functions *P*.

$$\Delta B_{z} = 0$$

$$\frac{B_{z}(r, \vartheta, \varphi)}{B_{0}} = 1 + \sum_{n=1}^{\infty} \left(\frac{r}{r_{0}}\right)^{n} \begin{bmatrix} H_{n} P_{n}(\cos \vartheta) + \\ \sum_{m=1}^{n} \left(I_{n}^{m} \cos m\varphi + J_{n}^{m} \sin m\varphi\right) W_{n}^{m} P_{n}^{m}(\cos \vartheta) \end{bmatrix}$$

$$|W_{n}^{m} P_{n}^{m}(\cos \vartheta)| \leq 1$$

$$H_{n}, I_{n}^{m}, J_{n}^{m} \propto \left(\frac{r_{0}}{a_{1}}\right)^{n}$$
Courtesy Pr. Guy Aubert
Unique set of coefficients $\rightarrow \vec{B}, \vec{A}, V^{*}, \vec{\Theta}$

Optimization of the homogeneity: cancel H_n, I_n^m, J_n^m



Set of coils of axe Oz, with a rectangular section and an uniform current density. Symmetry with respect to the xOy plan $\rightarrow H_{2p+1}=0$ and I, J=0

 $\frac{B_{z}(r, \mathcal{G}, \varphi)}{B_{0}} = 1 + \sum_{p=1}^{\infty} \left(\frac{r}{r_{0}}\right)^{2p} H_{2p} P_{2p}(\cos \mathcal{G})$ Minimize the coil volume for a given B₀ with H₂=H₄=...=H_{2p}=0 \rightarrow the non homogeneity is driven by the term H_{2(p_0+1)}



Need of at least p_0+1 coils to realize homogenous magnet at the 2(p_0+1) order « shimming theorem»

 \rightarrow Impossibility to cancel H₂ with only one winding of rectangular section

Courtesy Pr. Guy Aubert
$$H_2 \propto \left[\frac{b(a^2 + ac + c^2)}{c^3(a + c)}\right]_{a_1}^{a_2}$$
 a : radius
 $c = \sqrt{a^2 + b^2}$

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HELMHOLTZ COILS



$$B(z) = \frac{\mu_0}{2} \left[\frac{R^2}{\left[R^2 + (z+d)^2 \right]^{3/2}} + \frac{R^2}{\left[R^2 + (z-d)^2 \right]^{3/2}} \right]$$



Cancellation of the first two on-axis coeff. of the SHE







ERRORS AND IMPERFECTIONS

There are two main families of errors:

o **Errors that respect the basic symmetry** They are "built-in" in the design or they come from systematic manufacturing errors (coils systematically too large,...)

 \rightarrow Creates H_{2p} terms

o **Errors out of the basic symmetry** Scattering in material tolerances, in manufacturing and assembly,...

 \rightarrow Creates H_{2D+1} ; I, J terms (difficult to correct)



Need for compensation and shim coils to achieve the required homogeneity



Improve uniformity from ~500 ppm in magnet as-built to 10 ppm

Passive shimming

- Precisely-positioned pieces of iron in the warm bore. Improves overall uniformity (not individual harmonics)
- Lowest cost option
- <u>Cons:</u> Good for specified field/location only. Re-shim requires Service. Temperature drift

Active superconducting shimming (SC and resistive shims)

- Adjustable. Trade off convenience vs performance. Compensates harmonics
- <u>Cons:</u> Higher cost. Performance limitations. Need re-shimming if moved to a different location or even re-ramped. Interactions with magnet

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SC MRI MAGNETS



High magnetic field, high current, large useful volume, large stored energy, high mechanical forces and stresses

SC state requires low temperature

Complex cryogenic system; have to be optimized (compact, autonomous, minimum consumption)

Protection in case of quench

- Where to dissipate the stored energy?
- Manage the quick temperature elevation in the SC system
- Manage the large stresses
- Protection of the patient!!!

Advanced manufacturing techniques required

- Superconductor
- Winding (layer or DP, manufacturing accuracy and tight assembly tolerances)
- Electrical insulation



Reliable operation at customer sites (hospitals)

High field homogeneity

Multi-coils configuration Precise coil positions

Persistent operation

Superconducting joints

Switch

Passive quench protection, no external discharge

Minimized stray field

Shielding coils with reversed polarity

Passive cryogenics design Zero Boil Off

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MAGNETIC COMPONENTS OF A MRI SYSTEM



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Jeng in LTS and HTS conductors at 4.2K and 1.9K



Conductor Source: http://fs.magnet.fsu.edu/~lee/plot/plot.htm



A LARGE CHOICE OF SC WIRES AND CABLES...

MRI is biggest user of NbTi SC wire

NbTi

- Dominant commercial superconductor
- Bendable, ductile, low cost (\$1/kA.m)
- Tc=9,3K, Bc2=11,4 @ 4,23K

Nb3Sn

- Primary high field SC
- Brittle
- Tc=18K, Bc2 ≈ 23-29K
- Higher cost (x 5 price of NbTi)

MgB2

- Brittle
- Tc=39K, Bc2=40T
- Higher cost (x 5 price of NbTi)

Technology based on ReBCO super expensive (\$50-100/kA.m) and not mature enough for large industrial applications











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CRYOGENIC DESIGN





SC MRI MAGNET COOLING

- Volume production of MRI systems resulted in dramatic improvements in cryogenic performance
- 2000s: zero-boil-off (ZBO), or better, zero-helium loss refrigeration became standard in commercial MRI
 - Cryocooler re-condenses He gas inside the cryostat → No need in helium refill
 - Compact magnet design: one thermal shield eliminated
 - Disadvantages of ZBO:
 - Higher refrigeration cost
 - Higher power consumption

2010s: low-cryogen/no-cryogen technology

- GE *Freelium*™, Philips *Blue Seal*
- Less than 100 liters of LHe (ZBO: ~1,000 liters)
- Higher capital cost, not in volume production

	1980s	1990s	2000-10s	2020s
Technology	Nitrogen shield	GM cryocooler	ZBO	Low He
LHe boil off, cc/hr	0.4	$0.1 \Rightarrow 0.03$	Zero	Zero
LHe refill period	4 months	1 year ⇒ 4 years	Typically, no refill	No refill: closed system
LN refill period	1-2 weeks	Not used	Not used	Not used

Principal schematic of ZBO refrigeration



Courtesy M. Parish

QUENCH AND MAGNET PROTECTION

Challenges

- Unacceptable:
 - Safety risk in any mode
 - Magnet damage
- High energy
- High energy density
- External dump: unfeasible
- Emergency ramp-down
- High voltages and temperatures
- Interaction with other components
 - False protection activation
 - Multiple operation modes

Approach to MRI/NMR quench protection:

- Passive detection: no external circuit
- Heaters:
 - Resistive: use magnet energy
 - Inductive heating (ancillary)
- Energy released inside cryostat

System requirement: FOV field must be within ±0.05% peak-peak during years of the scanner operation



Two options:

- 1. Driven operation
 - Very stable power supply (expensive!!!), permanently installed
 - Lead losses

2. Persistent operation: average decay <0.1 ppm/hr

- Typical MRI/NMR configuration
- Lower cost. Better performance
- Energization is not possible on site
- Requires very low circuit resistance
- Retractable leads





SC JOINTS – A MAJOR TECHNICAL CHALLENGE

Challenges (NbTi joints)

- Even distribution of current in filaments
- Minimize field on joints. Consider field orientation
- Joints to conductor with different filament diameters
- Target resistance <10⁻¹¹ ohm per joint, lower for low-current units



US Pat. 3,422,529 (J.M. Nuding, 1969)

- Twisting required for good joint
- Should have a least 3 twists
- Clean in nitric acid
- Cylinder can be same superconducting material (Nb 25%Zr) or stainless steel
- Cold press at ~200 ksi

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MAGNET MECHANICAL ARCHITECTURE



- Coil former support coil during winding and coil-to-coil interaction forces
- Rods suspend cold mass to vacuum vessel and thermal shield to cold mass

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MRI AND SAFETY



MRI scanners are safe, if properly used and maintained

•No long-term negative effects of MRI scanning have been reported

•Safety and health hazards

- Static magnetic field
 - Ferromagnetic objects, implants
 - Biological effects, typically, short term
 - No known long-term effect at up to 8T.
 - First data at 10,5T, no negative feedback so far
- **o** Time varying magnetic field
 - Peripheral nerve and muscle simulation
- Pulsed radiofrequency field
 - Body heating, heat stress
- Acoustic noise
- Cryogens
 - No hazard to patients

SAFETY AND MAGNETIC FORCES

Magnetic objects in the scan room are prohibited!!!



Wheelchair

Blood trans-

fusion stand

- The force on magnetic object is proportional to the field gradient
- The gradient is the highest at the entrance to the magnet bore

 Magnetic force may be up to hundred times of the weight

• Implants, pacemakers



Floor polisher







MRI BUSINESS AND COMMERCIAL MRI MAGNET EXAMPLES

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MRI: THE LARGEST APPLICATION OF SUPERCONDUCTIVITY

MRI + NMR industries uses about 70% of all NbTi conductor



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MRI: THE LARGEST USAGE OF HELIUM



- Cryogenic applications use over 50M m³ of helium per year
- Zero-boil-off (ZBO) MRI magnets:
 - \circ Helium usage is reduced by about 75%
 - Most scanners do not require helium refill
 - Typical non-ZBO: >3 years between refills
- Challenges:
 - Insufficient demand
 - 2019-2022: higher price/reduced availability





MRI MARKET: LARGE AND GROWING

More than 50,000 MRI units of different types are installed worldwide

- Annual production: about 4,000 scanners
- ➤ Superconducting MRI: >75% of the installed base
- About 100 million MRI exams per year worldwide





EARLY NMR IMAGING MAGNETS 1977 - 1981



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Aberdeen 0.03T resistive magnet - 1977 0



0.15T Resistive magnet - 1980



John Woodgate and the first 0.3Tesla NMR Imaging Magnet for EMI



First 1.5Tesla magnet (STAR)

Courtesy G. Gilgrass

FAST EVOLUTION : X 5 WITHIN 20 YEARS FOR ACTIVELY SHIELDED MAGNETS



1985 - First 1.5 Tesla Active Shield Test Bed



2015: 7 Tesla Active Shield *First clinical* (FDA approved) system – first installations:

- University of Erlangen, Germany
- Cambridge University, UK.







Lighter and more compact!



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COMMERCIAL 3 T SUPERCONDUCTING SCANNERS WITH IMPROVED PATIENT ACCESS











MRI MAGNETS SHAPE

Cylindrical magnets: >99% of superconducting scanners



Open magnet

1 T Panorama (Philips)



Hybrid 0.5 T OpenSky MRI (Paramed)







Extremity scanner GE Optima 430s



ULTRA HIGH FIELD MRI MAGNETS

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SNR gain $\propto B_0$





3T

SNR~B₀^{1.65}

Pohmann et al. Magn Reson Med 2016;75:801–809

Improvement of spatial and temporal resolution



7T





WB MRI MAGNETS TYPICAL SIZE



Field	1,5 T	3 T	7 T	10,5T	11,75 T
	GE-SHFJ/CEA	Siemens	Siemens	Minneapolis	lseult/ Neurospin
Length (m)	1,25 - 1,7	1,6 - 1,8	~ 3	4,1	4
Diameter (m)	1,9 - 2,1	1,90 - 2,1	> 2,50	3,2	4,6
Mass (tons)	~ 5	~ 8	~ 25	~110	~ 135

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IMAGE QUALITY VS. MAGNETIC FIELD





1 to 2 mm resolution

 $7T \approx 0.3 \text{ mm}$ resolution

Van der Kolk et al. Euro J Radiol 2013; 82: 708-718

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WORLD UHF MRI PARK 2001-2020







- Installed in research sites and for investigational use
- About 70 7T whole-body scanners are in operation
- High image quality. Spatial resolution: ~0.5 mm
- Very heavy, very expensive, customized
- Require large rooms. Stray field 7 x 4 m (shielded), 9 x 5 m (unshielded units)
- Whole-body scanners (82 90-cm clear bore):
 - Length >3 m
 - Magnet weight >25 tons
 - Require 50 tons (actively shielded) to 400 tons of wall shielding



GE - University of Iowa (2014)



Siemens Terra (2015) Compact 7T, Length=270 cm



Siemens, NYU Len=3.3m, H = 3.6 m



Philips, Spinoza Center for Neuroimaging, Amsterdam

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10.5T WB MRI- FIRST IMAGE ON HUMAN BRAIN - 2020

Center for Magnetic Resonance Research, Department of Radiology, Minneapolis



Mass	110 tons
Central bore diameter	88 cm
Length	4.1m
Width	3.2m
Conductor	NbTi (433km)
Operating temperature	ЗK

Passive shielding





Magnetic Resonance in Medecine, Volume 84, Issue 1, July 2020



Two 11,7T WB magnets have been manufactured by ASG (Italy) for NIH (Bethesda, USA) and for NRI (Gachon University, Korea)





- Nominal current reached in the factory
- On site commissioning on-going

	NIH/NRI	Iseult	
Nominal current	246 A	1483 A	
Inner diameter	68 cm	90 cm	
Outer diameter	2.7m	5m	
Length	4m	5.2m	
Shielding	Passive	Active	
Mass	820 tons (magnet: 70 tons iron: 750 tons)	132 tons	
Operation mode	Persistent	Driven-mode	
Temperature	2.3K (saturated)	1.8K (superfluide)	
Helium bath pressure	64 mbar	1.2 bar	
Helium volume	3000L	7000L	
Stored energy	194 MJ	338 MJ	
Inductance	6400 H	308 H 49	

THE ISEULT 11.7 T MRI PROJECT

- B0 / Aperture
- 11.75 T / 900 mm
- Field stability
- Homogeneity
- 0.05 ppm/h
- < 0.5 ppm on 22 cm DSV

Stored Energy	338 MJ
Inductance	308 H
Current	1483 A
Length	5.2 m
Diameter	5 m
Weight	132 t

Magnet parameters



Neurospin Center CEA Saclay, France





Innovative solutions for a MRI magnet

- 170 NbTi double pancakes for the main coil
- 2 NbTi shielding coils to reduce the fringe field
- Cryostat for superfluid helium at 1.8 K, 1.25 bars
- Dedicated cryorefrigerator (80 l/h + 40 W @ 4.2 K)
- **Driven mode operation**, with two 1500 A power supplies for redundancy



11.7 T magnet section : in orange the windings, in blue the mechanical structure at1.8 K and in violet the cryostat



A DEDICATED COMPLEX INSTALLATION TO OPERATE THE MAGNET



Power supplies



Cryo-lines



48 V Batteries





Vacuum circuit



MCS/MSS/DAQ



Dump resistor

Control room

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INNOVATIVE DESIGN OF DOUBLE PANCAKE

$$B_{z}(r,\theta,\varphi) = B_{0} + \sum_{n=1}^{\infty} r^{n} \left[Z_{n}P_{n}(\cos\theta) + \sum_{m=1}^{n} \left(\begin{array}{c} \sum_{m=1}^{m} \cos m\varphi \\ + \\ \sum_{m=1}^{m} \sin m\varphi \end{array} \right) W_{n}^{m}P_{n}^{m}(\cos\theta) \right]$$



Magnet is theoretically intrinsically homogeneous

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COLD MASS STRUCTURE



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THE ISEULT MAGNET FABRICATION (2010-2017)



DP winding



Main coil - DP stacking



Shielding coil fabrication



MLI wrapping



Main coil / Shielding coils integration



Vacuum vessel welding



Magnet workshop, Belfort (France)

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2 WEEKS OF TRANSPORT FROM BELFORT TO SACLAY



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MAGNET INSTALLATION IN ITS ARCH









CALODUC ASSEMBLY – OCT 2017- OCT 2018

Caloduc : « umbilical cord » connecting the magnet with the cryogenic/electrical facilities



- « Helium tight » connection of cryogenic circuits, electrical connection of superconductors and of the inner magnet instrumentation : voltage taps (20), cryoshims (96), quench heaters (8), temperature sensors (96), strain gauges (24)
- **Complex mechanical structure** made of more than 200 parts (900kg mass)
 - Cold mass (1.8K)
 - Thermal Shield
 - Vacuum Vessel
 - Cooling circuits (TS and VV)
- **Mounting tolerances around 1 mm**; to take into account the thermal shrinkage during cool down





CALODUC ASSEMBLY – OCT 2017- OCT 2018



Helium vessel assembly



Dye penetrant tests



Cooling tubes local leak tests





Thermal shield



Vacuum vessel welding

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CALODUC COMPLETION - OCTOBER 2018





Final leak tests (inner cooling circuits and vacuum vessel)

Leak rate of 5.10-9 mbar.l/s on all the internal cooling circuits



COOLDOWN – 19 NOV 2018 – 7 MARCH 2019



- Huge mass to be cooled: cold mass (105 tons @ 1.8K) + thermal shield (3.4 tons @ 55K)
- Coolding rate limited by the thermal gradients across the coils (50K max)



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STEP BY STEP ENERGIZATION

Curre	ent					11.72T	11.72T
1500A]
1350A					10.551		. ا ۱۰۰۰ - ۱۰۰۰ - ۱۰۰۰ - ۱۰۰۰ - ۱۰۰۰ - ۱۰۰۰
1200A		· · · · · · · · ·		9.51T			
1050A			8.31				
900A		· · · ·	7T				
750A		5.9	T	· · · · ·	· · ·		
600A		· · · · ·		· · · · ·	· · · · · · · · · · · · · · · · · · ·	· · · · · ·	
450A		2	· · · · · · ·	· · · · ·	· · · ·	{· · · · · ·	
300A			· · · · · ·	· · · · ·			
1504	1.5 T						
1.50A							
UA	March 21, 2019				July 4 Ju	uly 18	Sept. 18, 2019



11.72T – JULY 18TH 2019

Test duration of 2 days

- Ramp-up in 30 hours
- Switching test between the two power supplies
- Plateau of 18 hours @ 11.72T
- · First magnetic measurements (homogeneity and drift)
- Slow discharge in 3 hours to unload the magnet







FIELD HOMOGENEITY MEASUREMENTS

• Magnetic measurements @ 1.5T, 3T, 7T and 11.72T



Field camera "499MHz"

- 40 NMR probes
- 50 cm diameter



Mechanical system designed to allow translations/rotations of the probe array

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FIELD HOMOGENEITY MEASUREMENTS

	300K	1.8K 1.5T	1.8K 3T	1.8K 7T	1.8K 11.7T	Cryoshim power
Z ₁ [PPM]	-132	-16	-7	-5	9	+/- 300
Z ₂ [PPM]	-105	-22	-16	-15	-17	+/- 70
Z ₃ [PPM]	20	2	2	2	2	+/- 10
X ₁ ¹ [PPM]	-1	22	21	22	24	+/- 32
Y ₁ ¹ [PPM]	59	63	59	60	62	+/- 32
X ₂ ¹ [PPM]	-	2	2	2	2	+/- 14
Y ₂ ¹ [PPM]	-	-6	-6	-6	-7	+/- 14
X ₂ ² [PPM]	-	-1	-1	0	-1	+/- 10
Y ₂ ² [PPM]	-	0	-1	-1	-1	+/- 10

- Very good agreement between the measurements
- Improvement of the field quality due to the shielding coils re-centering (behaviour anticipated during the design phase to specify the coil geometry at 300K)
 - Effect of thermal contraction at low temperature
 - Effect of magnetic forces that tend to push the two shielding coils toward the magnetic centre
 - Field homogeneity specification achievable using the available shimming power (cryoshim + iron shims), except for Y11
 - Additional passive shimming system for Y11 is being designed

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FIELD STABILITY



- Magnetic field drift adjusted using a fault current limiter
- 0.04 ppm/h obtained after only 4 hours of tests @ 11.72T (vs. spec 0.05 ppm/hour)
- Final setup will have to done for at least one week

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MRI EQUIPMENT INSTALLATION



Faraday cages



Gradient coil



Patient bed



FINAL COMMISSIONING STEPS

- Final magnetic field shimming
- Completion of the DAQ, control and protection systems and of the high availability equipment
- Installation of the Siemens MRI equipment
- New energization step by step
 - Adjustment of the field homogeneity (iron shims cryoshim) / stability
 - Commissioning of the high-availability control and protection systems
 - Tests of the GC vs. magnet interaction
 - Final ramp-up to 11.75T and final shimming

2020-2021 : First images of fruits, phantoms to setup the Iseult MRI system....



After 20 years, the brain first image!!!



 Commercial MRI magnets have reached maturity Efficient, well-integrated magnet design But there are still opportunities for improvement and growth

Superconducting MRI scanners
 The largest commercial application of superconductivity
 Dominate the marketplace

- NbTi is the conductor of choice for SC MRI magnets
- UHF MRI magnets are very promising but still remain the exception



Thank you for your attention

And thanks to Guy Aubert, Denis Le Bihan, Pierre Védrine, Thierry Schild, Alexandre Vignaux, Graham Gilgrass, Michael Parizh, Loris Scola, and Cécile Lerman

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