AAPM 2005 - Continuing Education Course - MRI Physics and Technology - 4

Advanced MRI - An Overview of Techniques and Applications

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Introduction

Image contrast in MRI depends on an extensive list of *intrinsic* and *extrinsic* parameters.

– Intrinsic parameters include:

- *proton density*
- *spin-lattice relaxation time (T_1)*
- *spin-spin relaxation time (T_2)*
- *chemical environment*
- *velocity*
- *diffusion*
- *perfusion*
- *temperature*

– Extrinsic parameters include:

- *echo time (TE)*
- *repetition time (TR)*
- *flip angle (α)*
- *contrast agents*
- *saturation pulses*
- *inversion pulses*
- *flow compensation pulses (GMN)*
- *diffusion sensitization pulses*

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Introduction

Four "advanced" MR applications to be discussed in this course:

– Assessing changes in ^{1}H diffusion
  - Use of diffusion imaging techniques to determine the rate and principle direction of thermal (Brownian) motion of protons.

– Assessing biochemical changes non-invasively
  - Use of MR spectroscopy (MRS).

– Assessing the microvascular environment
  - Use of dynamic contrast enhanced (DCE) or dynamic susceptibility change (DSC) MRI to assess changes in the microvascular environment.

– Assessing areas of neuronal activation
  - Use of blood oxygen level dependent (BOLD) MRI to determine regions of neuronal activation based on hemodynamic response.

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Assessing changes in ^{1}H diffusion

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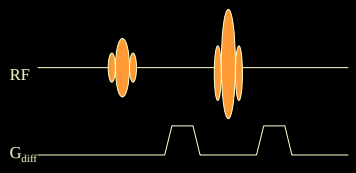
Diffusion imaging

The addition of two "diffusion-sensitizing" gradients provides a means of generating diffusion-weighted images.



In the presence of these gradients, the signal is attenuated according to $S/S_0 = e^{-bD}$, where D is the diffusion coefficient (mm^2/s), and *b* is the "*b*-value".

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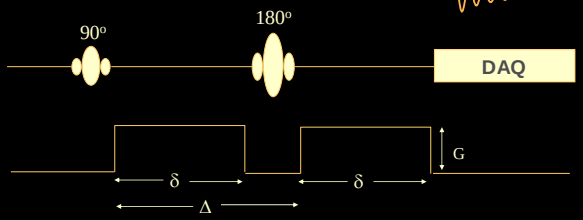
Diffusion imaging



$$b = \gamma^2 G^2 \delta^2 \left(\Delta - \frac{\delta}{3} \right)$$

Stejskal, Tanner. J Chem Physics , 1965

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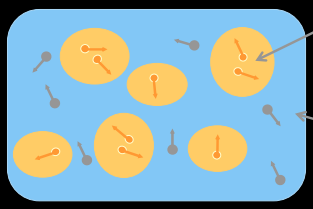
Diffusion imaging



$$D_{measured} = \frac{D_{intra}V_{intra} + D_{extra}V_{extra}}{V_{intra} + V_{extra}}$$

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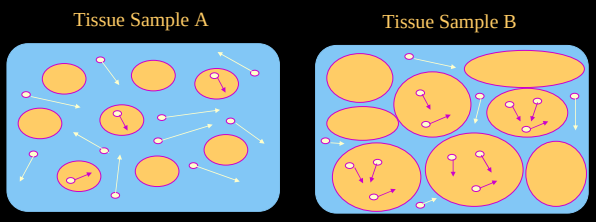
Diffusion imaging



Freely Diffusing Water = Dark

Restricted Diffusion = Bright

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Diffusion imaging

- Diffusion is described by a tensor.
- In materials with isotropic diffusion, the off-diagonal elements of the tensor are zero.
- In non-isotropic diffusing materials, the off-diagonal elements are non-zero, but (ideally) are symmetric.
- By applying the diffusion sensitizing gradient directions appropriately, the tensor elements can be completely defined.

$$\bar{\bar{D}} = \begin{bmatrix} D_{xx} & D_{xy} & D_{xz} \\ D_{xy} & D_{yy} & D_{yz} \\ D_{xz} & D_{yz} & D_{zz} \end{bmatrix}$$

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Diffusion imaging

Diffusion Anisotropy

It should be noted that in many tissues, *e.g.*, white matter tracts, the diffusion rates are anisotropic since water diffuses along the tracts more freely than transverse to the tracts.

The effects of anisotropy can be removed by acquiring images with *x*-, *y*-, and *z*-diffusion-weighted images and computing the trace of the diffusion tensor.

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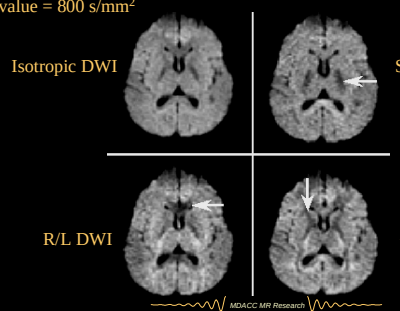
Diffusion imaging

b-value = 800 s/mm^2



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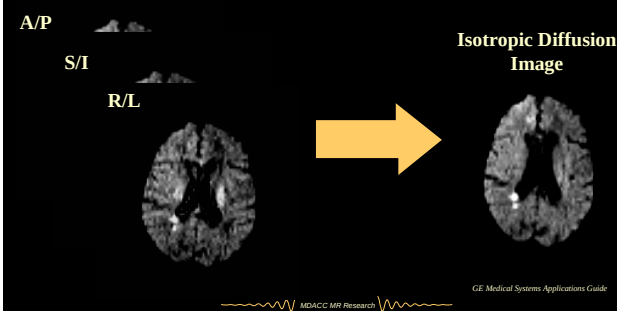
Diffusion imaging



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Diffusion imaging

Image Display

There are two ways of displaying images with diffusion-based contrast:

Diffusion-weighted imaging (DWI) - areas of unrestricted diffusion appear *hypo*intense and areas of restricted diffusion appear *hyper*intense.

Advantage: Very fast, no extra post-processing steps.

Disadvantage: T_2 "shine through" can be problematic. (Due to T_2 weighting of the sequence, *i.e.*, long TE intervals, necessary in order to obtain adequate *b*-values. This problem is decreased with high performance gradient subsystems.)

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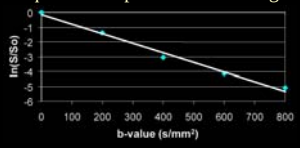
Diffusion imaging

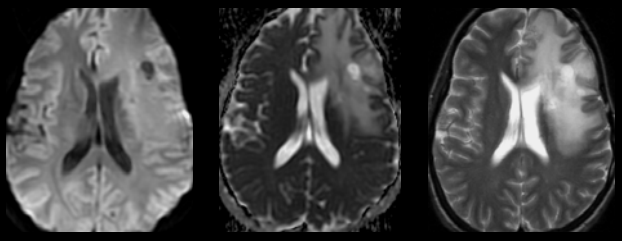
Apparent diffusion coefficient (ADC) imaging - areas of unrestricted diffusion appear *hyper*intense and areas of restricted diffusion appear *hypo*intense. Requires the acquisition of multiple sets of DWIs with varying *b*-values to allow computation of ADC values on a pixel-by-pixel basis by linear regression analysis of the signal attenuation equation, $ln(S/S_0) = -b * ADC$.

Advantage: T_2 "shine-through" is eliminated.

Disadvantage: Requires multiple *b*-values for regression => extra time.



ADC vs DWI



DWI | ADC (2 point) | T_2-weighted

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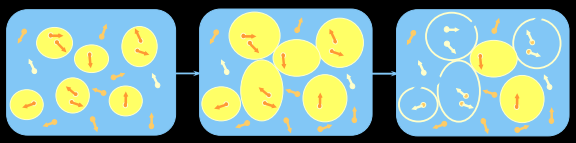
Diffusion imaging - Ischemic injury



normal tissue | cells swell | membranes break

D_{normal} | $D < D_{normal}$ | $D > D_{normal}$

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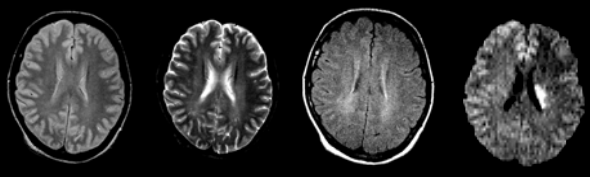
Diffusion imaging in acute stroke



PDW | T2W | FLAIR | Diffusion

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Diffusion weighted imaging

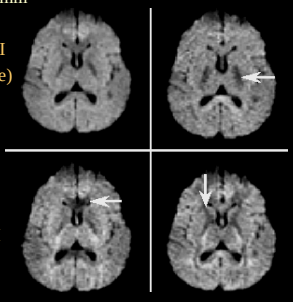
b-value = 800 s/mm²

Isotropic DWI ("Trace" image)

S/I DWI

R/L DWI

A/P DWI



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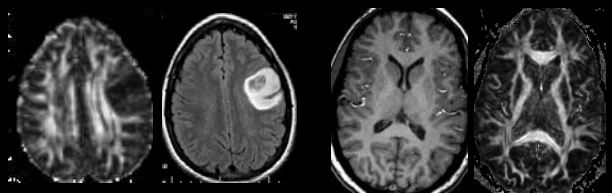
Diffusion tensor imaging (DTI)

Using multiple diffusion encoding directions to determine the diffusion tensor terms, eigenvalue/eigenvector analysis can be used to determine the *principle diffusion direction*. This is the basis of "tractography".



1.5T, b=1576 s/mm^2, 6 directions 3.0T, b=1000 s/mm^2, 15 directions

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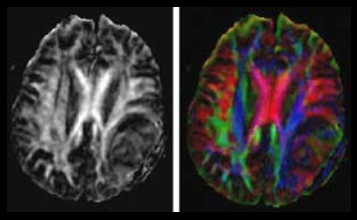
Diffusion tensor imaging (DTI)

Furthermore, fractional (or relative) anisotropy indices can be computed to more fully characterize the white matter tract directions.



Red: Right/left

Green: Anterior/posterior

Blue: Superior/inferior

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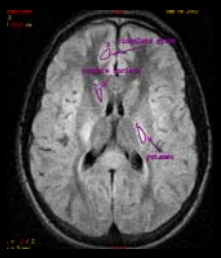
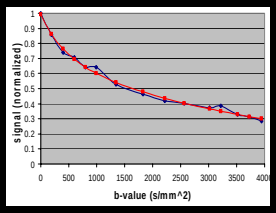
Image from X. Joe Zhou, Ph.D.

Cell volume fraction measures





$$S = 0.705e^{-b(0.0002)} + 0.295e^{-b(0.002)}$$

$$D_{measured} = \frac{D_{intra}V_{intra} + D_{extra}V_{extra}}{V_{intra} + V_{extra}}$$

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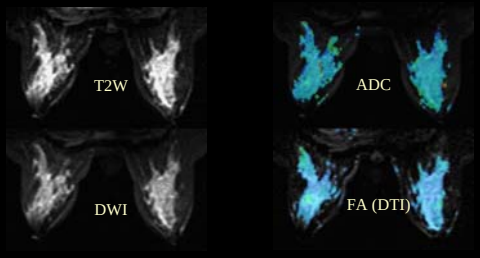
X. Joe Zhou, PhD, Rebecca Milman, MS

Breast diffusion MRI



Breast Diffusion Imaging @ 3T

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Assessing biochemical changes

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Introduction to spectroscopy

- The Larmor relation that relates resonant frequency to magnetic field strength:

$$\nu = \gamma \ B_{nucleus}$$

where ν is the resonant frequency (MHz), γ is the gyromagnetic ratio, and $B_{nucleus}$ is the magnetic field strength (T) *at a given nucleus*.

- However, the value of $B_{nucleus}$ depends on the local electronic environment, *i.e.*, it is the value of the applied field, B_o, modified by the magnetic field due to the chemical environment.

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Introduction to spectroscopy

- Nuclei in different chemical (electronic) environments will have slightly different resonance frequencies depending on the amount of local nuclear shielding, σ:

$$\nu = \gamma B_o (1 - \sigma)$$

- It is this local shielding effect that results in spectra with multiple peaks for a given nuclear species, where the peak positions depend on the local chemical environment.

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Introduction to spectroscopy

- The position of a given spectral peak is usually given in terms of *chemical shift* with respect to some reference,

$$\delta_x = (\nu_x - \nu_{ref}) / \nu_{ref}$$

- This definition makes the separations between the peaks independent of applied field strength.
- Note, however, that the separation of the peaks (in Hz) *does* depend on field strength. Therefore, the spectral resolution improves as field strength increases. (So does the SNR of the spectral peaks.)
- For ^{1}H MRS applications, the reference is usually water.

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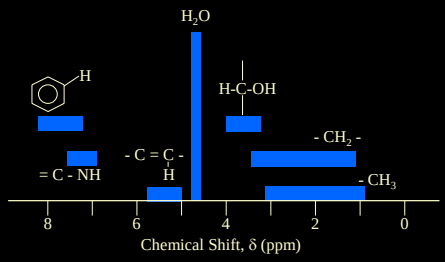
Introduction to spectroscopy



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Requirements

The success of an MRS examination depends upon the following:

- High quality localization
  The volume from which the chemical information (spectrum) is obtained must be accurately known.
- Highly homogeneous magnetic field
  Linewidths of peaks are inversely proportional to T_2*, so improved homogeneity results in narrower peaks (improved spectral resolution).
- Efficient water suppression (^{1}H MRS)
  This is aided by improved homogeneity as well.
- Spectral quantitation

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Water suppression (^{1}H MRS)

- The metabolites of interest are usually about a factor of 8,000 less in concentration than water.
- A very efficient means of suppressing the water resonance is required in order to readily detect the metabolite resonances.

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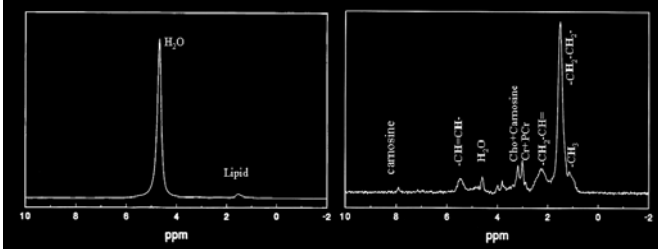
^{1}H MRS - Gastrocnemius muscle



Without H_2O Suppression

With H_2O Suppression

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Water suppression (1H MRS)

- The most commonly utilized method for water suppression is based on the same principle as “fat sat”. (For MRS sequences, the suppression pulses are commonly referred to as CHESS pulses - *chem*ically selective saturation.)
- Typically, multiple (often 3), narrow bandwidth (~50 Hz) pulses are applied at the water resonance frequency preceding the localization sequence.
- Multiple pulses are used to improve the degree of water suppression.

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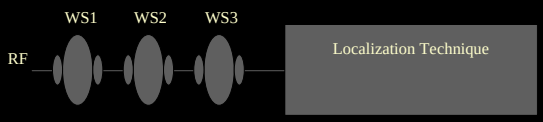
Water suppression (^{1}H MRS)



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Localization

For a spectrum to have any significance, the region from which it is obtained must be accurately known. The most commonly used localization techniques are:

- Single voxel localization (SVL): The VOI is the intersection of three slice selective gradient/RF pulses. Each slice thickness can be individually varied to define VOI.
- Spectroscopic imaging (SI): Uses phase-encoding for localization.
- Hybrids: Usually a combination of SVL and SI techniques.

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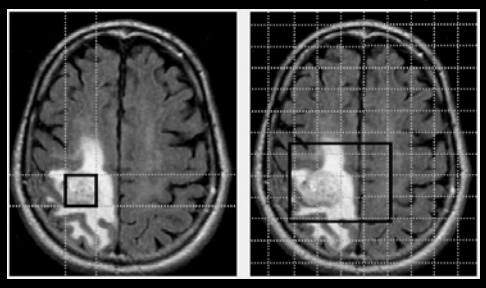
Localization techniques



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Single voxel localization

- The most common single volume localization techniques are those based on the *s*timulated *e*cho *a*cquisition *m*ode (STEAM) and *p*oint *re*solved *s*pectroscopy (PRESS) sequences.
  - STEAM: 90º-90º-90º-acquire
  - PRESS: 90º-180º-180º-acquire
- Advantage of STEAM: shorter minimum echo times
- Advantage of PRESS: 2x SNR increase compared to STEAM (for peaks with no *J*-coupling)

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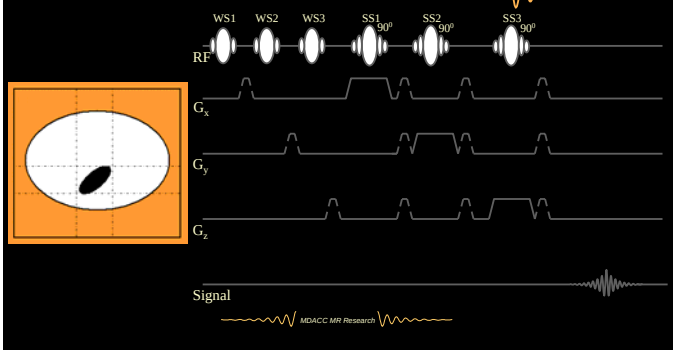
STEAM



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Spectroscopic imaging techniques

Instead of relying on the intersection of three planes to define a single VOI, SI techniques use phase-encoding for part or all of the localization to yield multiple VOIs.

- 2DSI: Uses one slice selection gradient/RF pair to define a slice, and then phase-encodes the remaining two dimensions. (Most commonly used SI method.)
- 3DSI: Uses three phase-encoding gradients to define a 3D volume of voxels.

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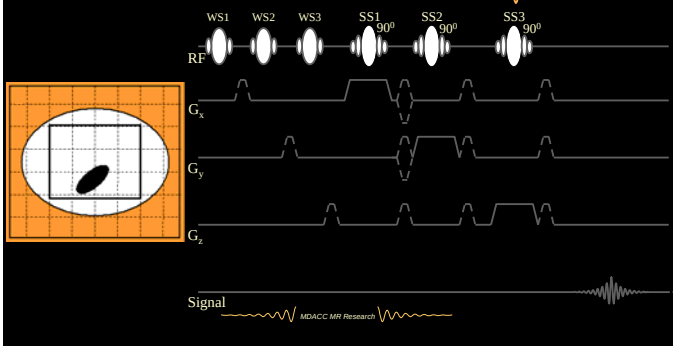
Spectroscopic imaging (SI)



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Spectroscopic imaging techniques

While SVL techniques are faster for obtaining a single localized spectrum, SI techniques have the following advantages:

- Spectra from multiple VOIs can be obtained for comparison.
  - Useful for comparing suspected pathological tissue with normal-appearing contralateral region, or for better assessment of lesion heterogeneity.
- Spectra from smaller VOIs can be obtained as compared to SVL techniques.
  - Less partial volume averaging, better assessment of heterogeneity.
- "Metabolite maps", in which pixel intensity is proportional to chemical concentration, can be generated.

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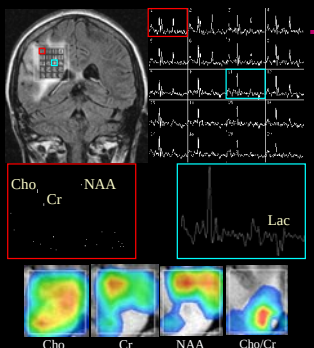
SI "Met Maps"



^{1}H MR Spectroscopic Imaging Applications

NAA:	*N*-acetylaspartate
Cho:	Choline compounds
Cr:	Creatine/ phosphocreatine
Lac:	Lactate

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Spectroscopic imaging techniques

Disadvantages of SI techniques include:

- rather long acquisition times:

  2DSI: $T_{scan} = N_{x\_phase} \times N_{y\_phase} \times N_{averages}$

- spatially-dependent water suppression efficiency & spectral quality
  - Larger volume over which field homogeneity must be optimized -- more difficult to accomplish than with SV localization.
- "spectral-bleed" from one voxel to another is possible due to phase-encoding point spread function. (Can be minimized by increasing N_{phases}, but this costs time.)

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What can be seen?

In ^{1}H MRS of the brain, the primary peaks are:

- NAA: *N*-acetylaspartate (viable neurons only) 2.0 ppm
- Cr: total creatine (creatine + phosphocreatine) 3.0 ppm
- Cho: choline (phosphotidylcholine, *etc.*) 3.2 ppm
- Lac: lactate 1.4 ppm

(All of the above can be detected at short *and* relatively long TE acquisitions.)

- GABA: γ-aminobutyric acid
- Simple amino acids, *e.g.*, alanine, glutamate, glutamine
- NAAG: *N*-acetylaspartylglutamate
- Asp: aspartate

(The above are in the 2.2-2.6 ppm range and typically require short TE acquisitions.)

- Lipids: range of chemical shifts, but dominant is methyl at 1.3 ppm
- Ins: *myo*-inositol 3.6 ppm
- Glucose (at ~ 3.5 ppm)

(The above require relatively short TE acquisitions.)

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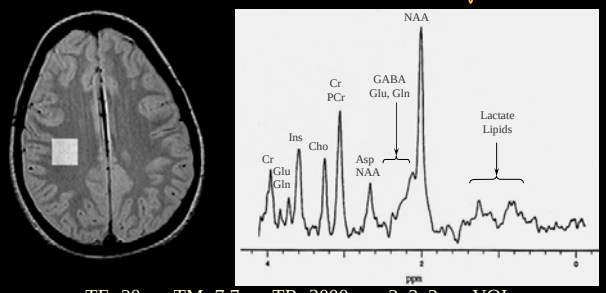
Short TE ^{1}H MRS



TE=20ms, TM=7.7ms, TR=3000ms - 2x2x2 cm VOI

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Choice of echo time

- As you increase TE, the signal intensity from each metabolite decreases due to spin dephasing.
- Short TE: more spectral peaks means improved chances for lesion characterization or assessment of therapy. However, the examinations are more difficult to obtain reproducibly, mainly due to decreased water suppression efficiency.
- When comparing MRS data, the TE and TR values must be taken into account.
- Each metabolite has its own T_1 and T_2 relaxation times. Therefore, as you change TE and/or TR, the relative areas and amplitudes change for each peak. Peak or area ratios are also TE-dependent.

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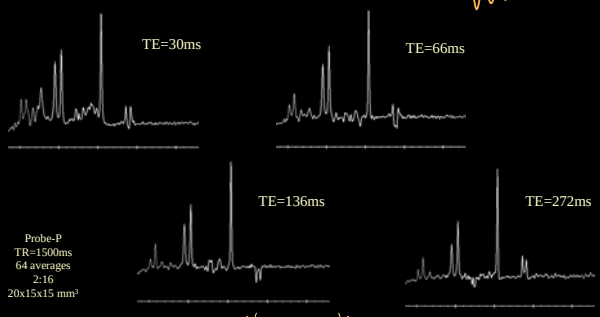
Choice of echo time



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Spectral quantitation

Quantitative analysis:

- Relative concentrations
  - Most commonly involves taking the ratio of peak areas or amplitudes, *e.g.*, NAA/Cr, Cho/Cr.
  - Problem: Changes in ratios can be due to changes in, for example, NAA *or* Cr.
- Absolute concentrations
  - Much more difficult and requires some form of "standard".
  - External standard: small container of known concentration of reference sample from which reference spectrum is obtained.
  - Internal standard: most commonly taken as water.

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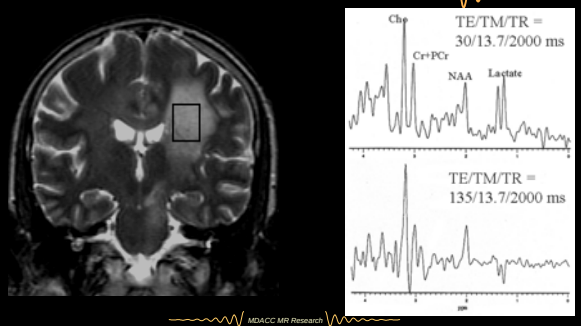
^{1}H MRS - Glioma



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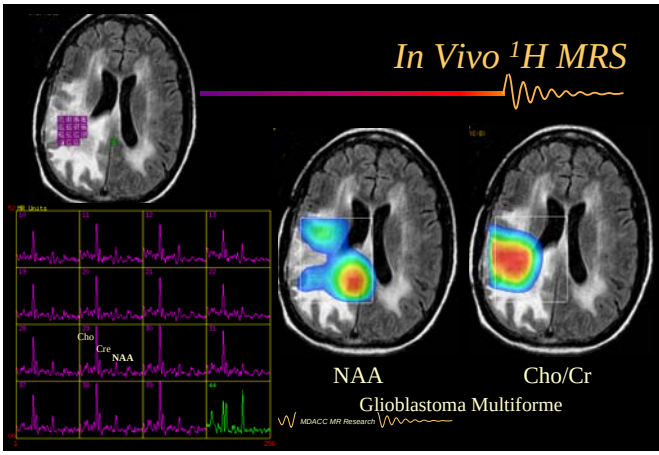
In Vivo ^{1}H MRS



NAA Cho/Cr

Glioblastoma Multiforme

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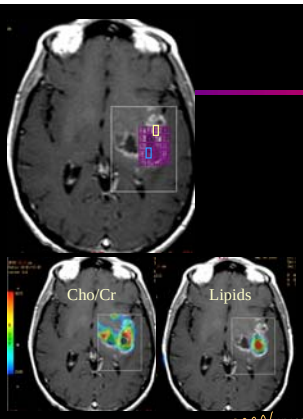
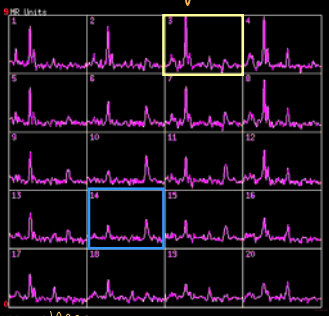
In Vivo ^1H MRS





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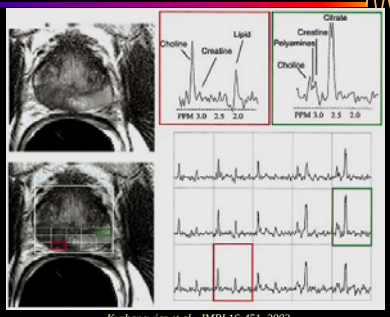
Prostate ^1H MRS



Kurhanewicz et al., JMRI 16:451, 2002

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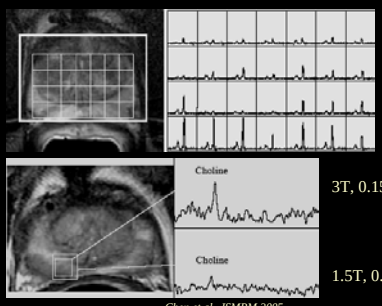
Prostate ^1H MRS



3T, 0.15 cm^3 voxels

1.5T, 0.30 cm^3 voxels

Chen et al., ISMRM 2005

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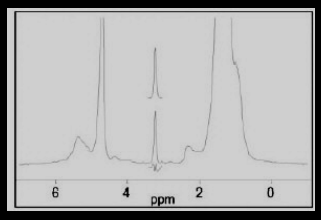
Breast ^1H MRS



Invasive Ductal Carcinoma

Bolan et al., MRM 50:1134, 2003

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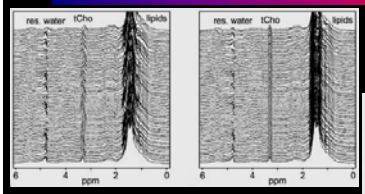
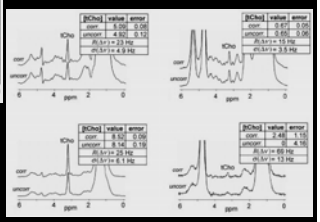
Breast ^1H MRS



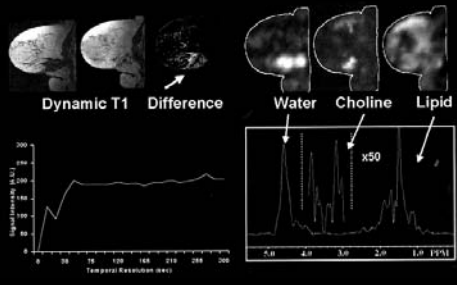


Frequency correction (cross-correlation) to compensate for respiratory motion.

Bolan et al., MRM 52:1239, 2004

MDACC MR Research

Breast ^1H MRS



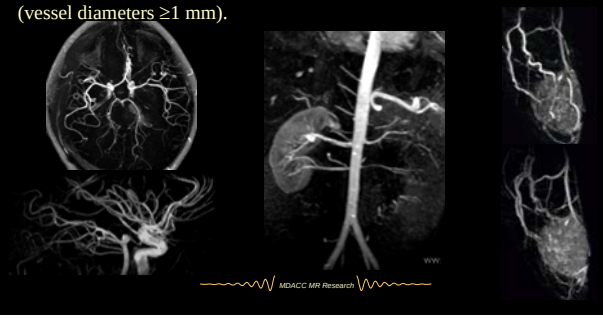
Jacobs et al., JMRI 21:23, 2005

MDACC MR Research

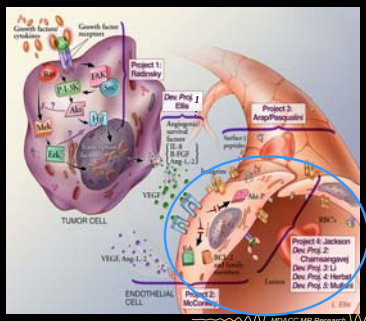
Assessing the microvascular environment

Assessing microvascular changes

MR angiographic techniques can assess macroscopic vascular morphology (vessel diameters ≥1 mm).



Assessing microvascular changes



Goal:

Non-invasive assessment of the effects of antiangiogenic / antivascular therapy.

Assessing microvascular changes

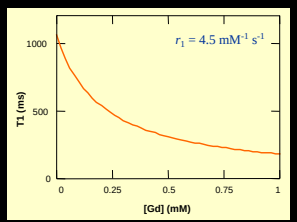
- The most common MR techniques for assessing microvascular changes:
  - Dynamic contrast agent enhanced MRI (DCE-MRI)
  - Dynamic susceptibility change MRI (DSC-MRI)
- Both require rapid temporal sampling, with preferred sampling rates on the order of
  - 5 - 10 sec per image set for DCE-MRI
  - 1 - 2 sec per image set for DSC-MRI
- Both require the infusion of exogenous contrast agents.

Common MRI contrast agents

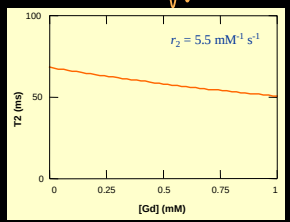
Paramagnetic Contrast Agents

- Gadolinium is the most common paramagnetic atom used in MR agents
- Gd is toxic - must be tightly chelated
- Three common Gd agents:
  - Magnevist (gadopentetate dimeglumine) ionic
  - Omniscan (gadodiamide) non-ionic
  - Prohance (gadoteridol) non-ionic
- Osmotic loads of all three are *significantly* less than iodinated agents
- Affect both T_1 and T_2 relaxation times, with the dominant effect being shortening of the T_1 relaxation time (at routine clinical doses).

Paramagnetic contrast agent effects



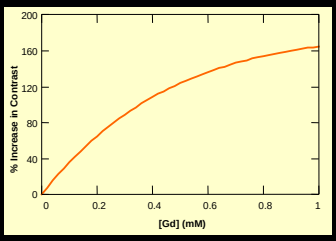
$$\frac{1}{T_1} = \frac{1}{T_{1,0}} + r_1[Gd]$$

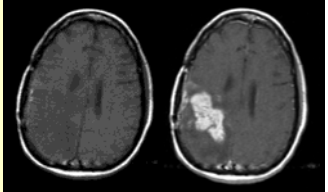


$$\frac{1}{T_2} = \frac{1}{T_{2,0}} + r_1[Gd]$$

Effects of increasing Gd-DTPA concentration on T_1 (left) and T_2 (right) relaxation times in gray matter ($T_{1,0}$ = 1055 ms, $T_{2,0}$ = 68 ms). Note the dominant effect on T_1 relaxation times.

Paramagnetic contrast agent effects





Percent increase in contrast for gray matter as a function of Gd-DTPA concentration for a SE sequence with TR/TE = 400ms/18ms.

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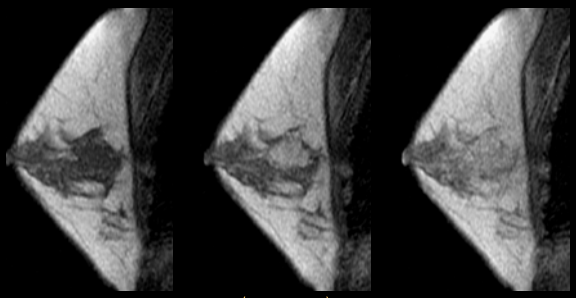
DCE-MRI - Background



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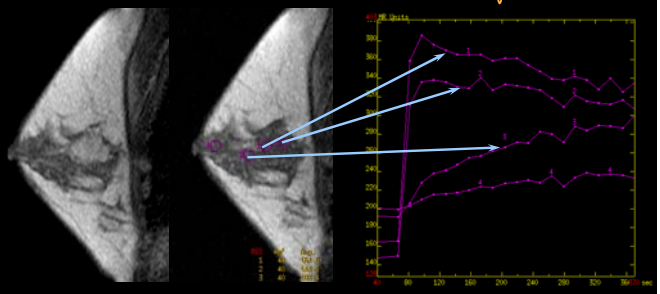
DCE-MRI



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DCE-MRI

- DCE-MRI acquisitions typically are based on fast spoiled gradient-echo sequences (FSPGR, FLASH):
  - Spoiling maintains T_1-weighting even with very short TRs
  - Trade-off between need for good temporal resolution and adequate spatial coverage
  - Both 2D and 3D acquisition modes have been used

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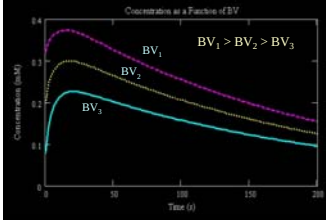
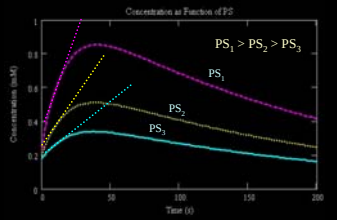
DCE-MRI analysis



Two-Compartment Pharmacokinetic Model (Blood Volume Effects)



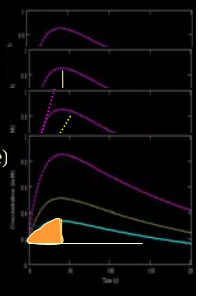
Two-Compartment Pharmacokinetic Model (Permeability or Extraction Flow Effects)

MDACC MR Research

DCE-MRI analysis

Choice of Analysis Methods:

- Qualitative
  - Visual examination of uptake curves
- Quantitative, but no physiological basis
  - Time to peak enhancement
  - Maximum uptake (maximum signal difference)
  - Maximum rate of uptake (maximum slope)
  - Area under the curve (AUC) and initial AUC
- Quantitative, with physiological basis
  - Pharmacokinetic modeling



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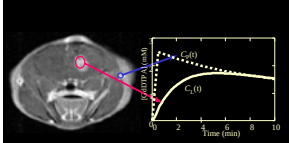
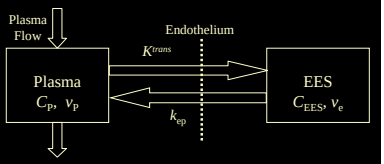
Two Compartment Pharmacokinetic Model





Measured Measured

$$C_L(t) = v_P C_P(t) + C_{EES}(t)$$

$$C_{EES}(t) = K^{trans} \int_0^t C_P(t')\ e^{-k_{ep}(t-t')}\ dt'$$

C_P = [Gd] in plasma (mM) = C_b / (1-Hct)

C_{EES} = [Gd] in extravascular, extracellular space (mM)

K^{trans} = endothelial transfer coefficient (min^{-1})

k_{ep} = reflux rate (min^{-1})

v_P = fractional plasma volume, v_e = fractional EES volume

Standardized parameters as proposed by Tofts *et al.*, *J Magn Reson Imaging*, 10:223-232, 1999.

MDACC MR Research

Quantitative methods

Quantitative DCE-MRI data analysis

- Primary model: two-compartment pharmacokinetic model
- Regimes:
  - Flow-limited (Kety model) - high permeability
  - Permeability surface area-limited - low permeability
  - Mixed model
- Measures:
  - K^{trans} - endothelial transfer coefficient (min^{-1})
  - v_P - fractional plasma volume
  - k_{ep} - reflux rate (min^{-1})

MDACC MR Research

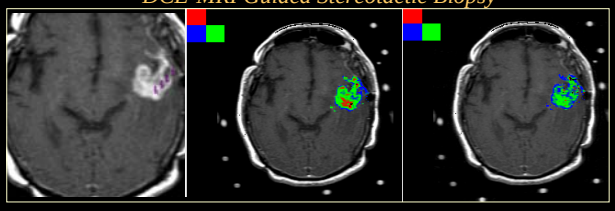
Quantitative methods

- Flow-limited case:
  - K^{trans} => EF --- the "extraction flow product"
    - $E = (1 - e^{-PS/F})$
    - Most generally true for current FDA-approved small MW agents!
- Permeability-limited case:
  - K^{trans} => EF => PS
    - Since E => PS / F
    - Typically true for contrast agents with MW > ~50 kD

MDACC MR Research

DCE-MRI - Parametric mapping

DCE-MRI Guided Stereotactic Biopsy



DCE-MRI T1W Image | Plasma Volume Fraction Map | K^{trans} Map

MDACC MR Research

PTK787/ZD222584 – Liver Mets

Dynamic Contrast-Enhanced Magnetic Resonance Imaging As a Biomarker for the Pharmacological Response of PTK787/ZK 222584, an Inhibitor of the Vascular Endothelial Growth Factor Receptor Tyrosine Kinases, in Patients With Advanced Colorectal Cancer and Liver Metastases: Results From Two Phase I Studies

By Bruno Morgan, Anne L. Thomas, Joachim Drevs, Juergen Hennig, Martin Buchert, Asvina Jivan, Mark A. Horsfield, Klauss Mross, Howard A. Ball, Lucy Lee, William Mietlowski, Stefan Fuxius, Clemens Unger, Ken O'Byrne, Andrew Henry, Graham R. Cherryman, Dirk Laurent, Margaret Dugan, Dieter Marmé, and William P. Steward

Phase I Trial
26 patients
Metastatic liver lesions
(Colorectal cancer)

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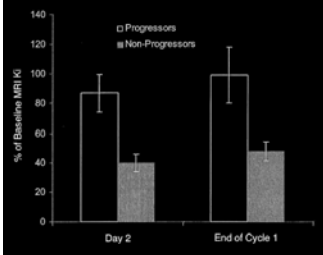
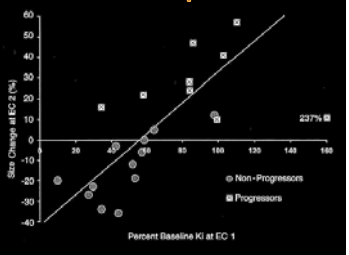
J Clin Oncol, 21:3955-64, 2003

PTK787/ZD222584 – Liver Mets





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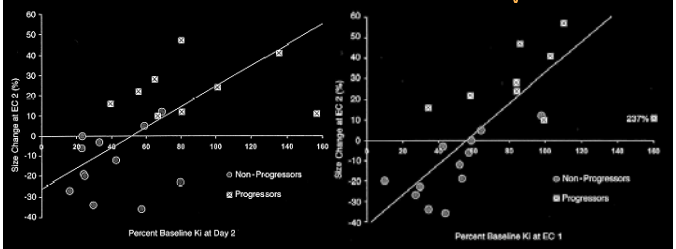
J Clin Oncol, 21:3955-64, 2003

PTK787/ZD222584 – Liver Mets



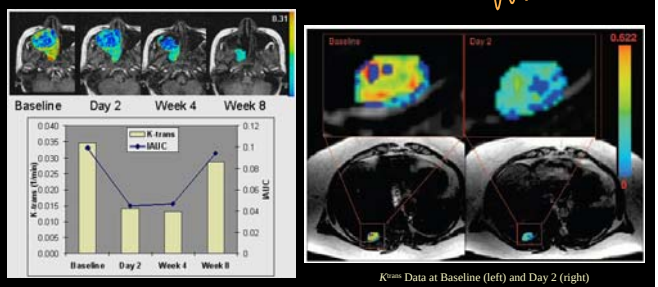
J Clin Oncol, 21:3955-64, 2003

AG-013736 DCE-MRI Study

AG-013736 Trial (DCE-MRI and DCE-CT)

- Potent and selective inhibitor of VEGFR/PDGFR tyrosine kinases
- Preclinical activity in xenograft models (melanoma, colon, breast, and lung)
- Multicenter Phase I study in solid tumors (MDACC, University of Wisconsin, UCSF)
- Heterogeneous lesions (liver, lung, head & neck, …)
- Data analyses performed at VirtualScopics, LLC (Rochester, NY) and independently at MDACC

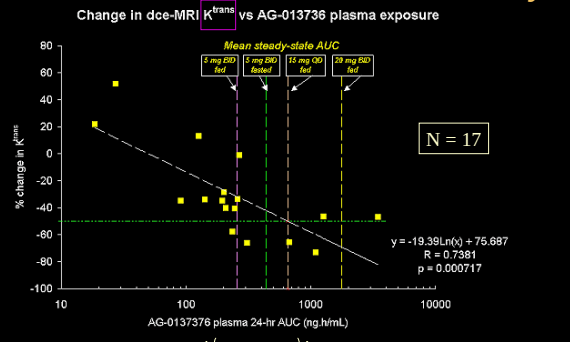
AG-013736 DCE-MRI Study



K^{trans} Data at Baseline (left) and Day 2 (right)

McShane, Ashton, Jackson, et al., Proceedings of the ISMRM, p 154, 2004

AG-013736 DCE-MRI Study

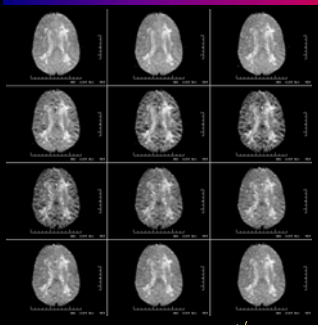


Liu et al., J Clin Oncol, in press.

DSC-MRI techniques

- Dynamic susceptibility change (DSC) MRI techniques have also been used to assess changes in regional blood flow.
- DSC-MRI uses T_2- or T_2*-weighted, high speed imaging techniques, *e.g.*, echo-planar imaging.

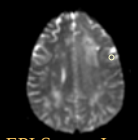
DSC-MRI principles



0.2 mmol/kg gadodiamide bolus infusion at 5 cc/sec

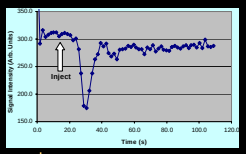
SE-EPI
TE/TR = 80/1700 ms
30 cm FOV, 128x128 matrix
125 kHz, 5 mm slice, 1.5 mm gap
65 phases, 1:52 min

DSC-MRI

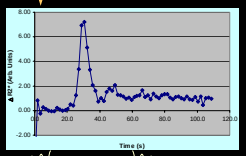


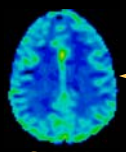
EPI Source Image

Extract $S(t)$



$$\Delta R_2^* = -1/TE \; ln[S(t)/S(0)]$$





rCBV Map

$$rCBV = \int_0^\tau \Delta R_2^*(t)\,dt$$

$$rMTT = \frac{\int_0^\tau \tau\,\Delta R_2^*(t)\,dt}{\int_0^\tau \Delta R_2^*(t)\,dt}$$

$$rCBF = \frac{rCBV}{rMTT}$$

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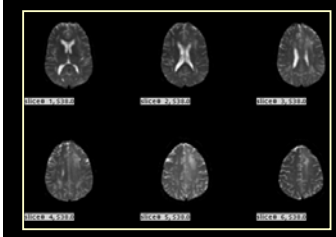
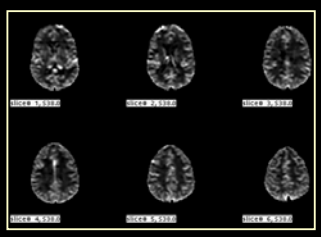
DSC-MRI



SE-EPI Source Images



rCBV Maps

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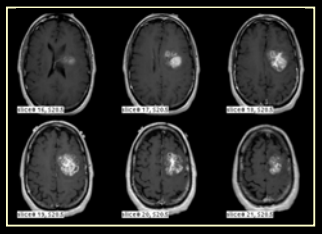
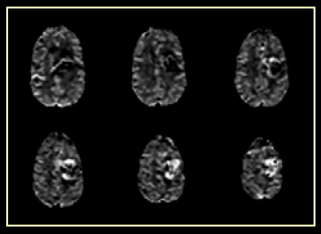
DSC-MRI



T_1-weighted Post-Gd



Computed rCBV Maps

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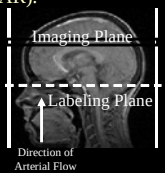
Other MR techniques to assess microvascular changes

- There are MR-based techniques for assessing changes in regional blood flow and/or volume.
- Arterial spin labeling (ASL) techniques do not require administration of exogenous contrast agents.
  - Are low SNR techniques, thus 3T systems will be beneficial.
  - Are associated with high specific absorption rates (SAR).
  - Are not yet commercially available.



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Assessing areas of neuronal activation

MDACC MR Research

BOLD functional MRI - Principles

*Blood oxygen level dependent **(BOLD)** contrast*

- Principle: Uses the difference in the magnetic state of oxyhemoglobin (diamagnetic) *vs* deoxyhemoglobin (paramagnetic) to provide image contrast.
- Advantage: Totally noninvasive. Requires no infusion.
- Disadvantage: Much smaller change in signal intensity compared to bolus injection technique (~1-5% changes at 1.5T).

References: Ogawa *et al.*, Magn Reson Med 14:68, 1990; Kwong *et al.*, Proc Natl Acad Sci USA 89:5675, 1992.

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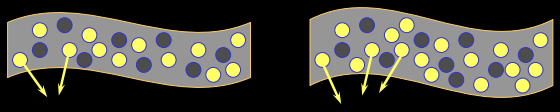
Functional MRI - BOLD Principles

*Blood oxygen level dependent **(BOLD)** contrast*



Normal State	*Neuronal Activation State*
• ~ 60% oxyHb	• ~ 75% oxyHb (↑ Flow & volume)
• Normal oxygen extraction rate	• ~ 5% ↑ in oxygen extraction rate
• Relatively large susceptibility effect	• Decreased susceptibility effect
• Baseline MR signal intensity	• Increased MR signal intensity

References: Ogawa *et al*., Magn Reson Med 14:68, 1990; Kwong *et al*., Proc Natl Acad Sci USA 89:5675, 1992.

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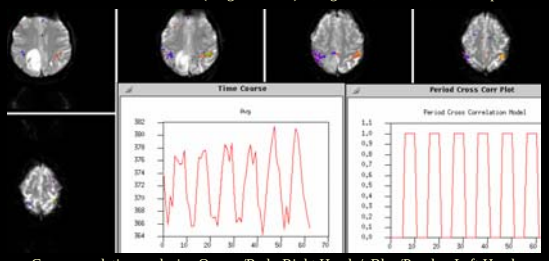
Functional MRI

Bilateral Motor Task Activation (Finger-Thumb) -- Right 15s, Left 15s, with 6 repetitions



Cross-correlation analysis: Orange/Red - Right Hand / Blue/Purple - Left Hand

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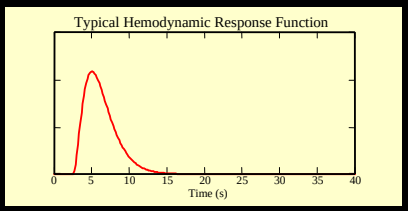
Hemodynamic response function



Approximate hemodynamic response function for delta function stimulus

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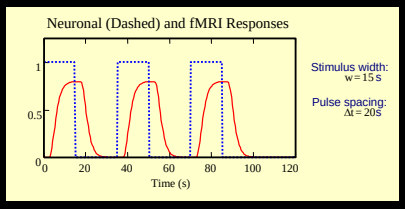
Latency in fMRI response



~2 sec delay from time of activation until start of fMRI signal response
~6-8 sec delay from time of activation until peak of fMRI signal response

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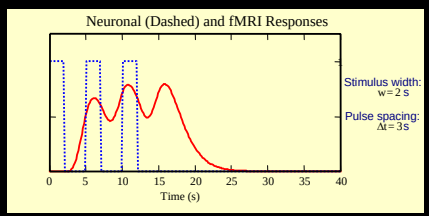
Latency in fMRI response



~2 sec delay from time of activation until start of fMRI signal response
~6-8 sec delay from time of activation until peak of fMRI signal response

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fMRI stimulation devices

MR-Compatible Video Goggles

MR-Compatible Audio Headphones

MR-Compatible Response Pads

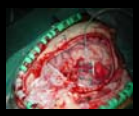
MDACC MR Research

fMRI applications in oncology

Initial Clinical Application

Neurosurgical planning

- Goal: Maximize resection volume (for best prognosis) while preserving "eloquent centers" (for quality of life).
- Benefits:
  - Pre-surgical planning
  - Decreased OR time
    - Replace evoked potential mapping (sensorimotor)
    - Minimize need for awake craniotomy / direct cortical stimulation (speech / memory)
  - Replace pre-surgical Wada procedure (speech / memory)



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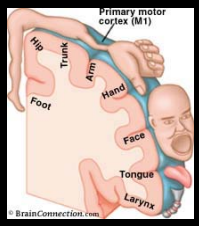
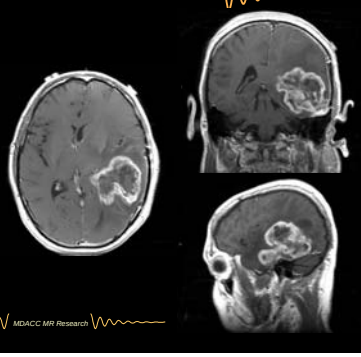
Locations of "Eloquent Centers"





Motor Homunculus

Ref: BrainConnection.com

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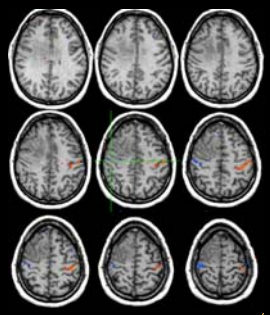
Functional MRI - Motor



Bilateral Hand Mapping for Presurgical Planning

Paradigm:

Bilateral finger-thumb tapping
15s off, (15 s right, 15 s left) x 5

Sequence:

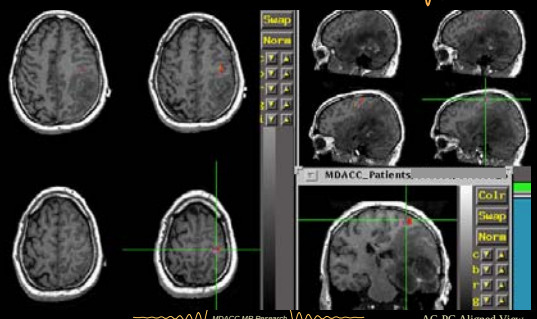
Single-shot GR-EPI
128 x 128 matrix, 32 x 32 cm FOV
6 sections, 55 phases, 2:45 min

Analysis:

Cross-correlation analysis, p=0.001, r=0.443.

MDACC MR Research

Functional MRI - Motor



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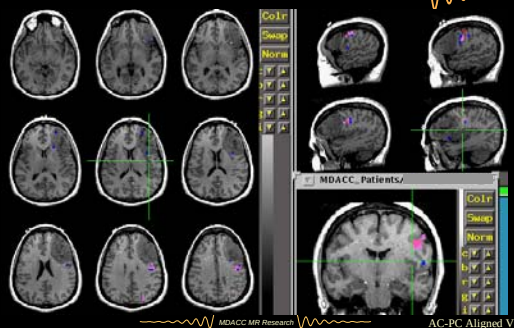
AC-PC Aligned View

Functional MRI - Expressive speech



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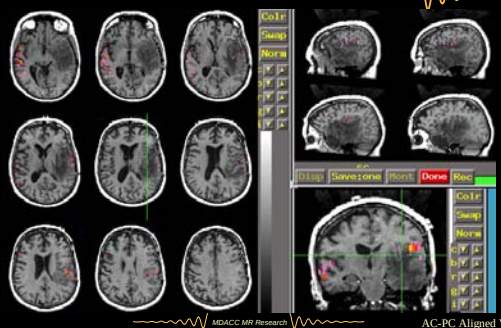
AC-PC Aligned View

Functional MRI - Receptive speech



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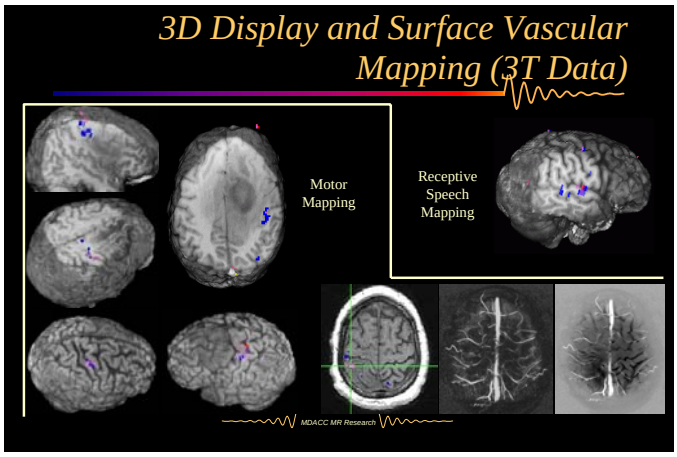
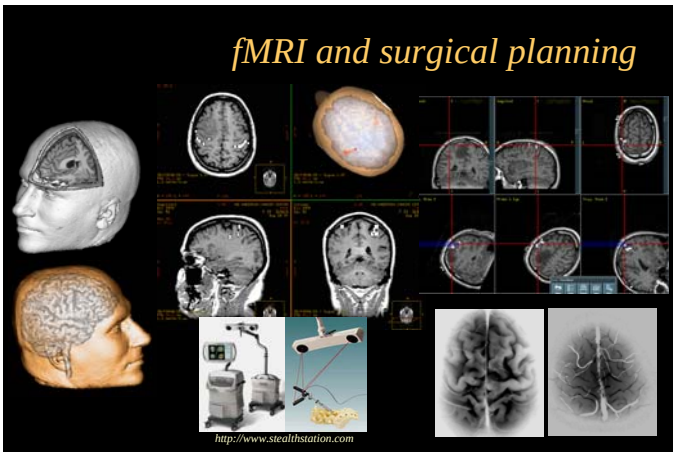
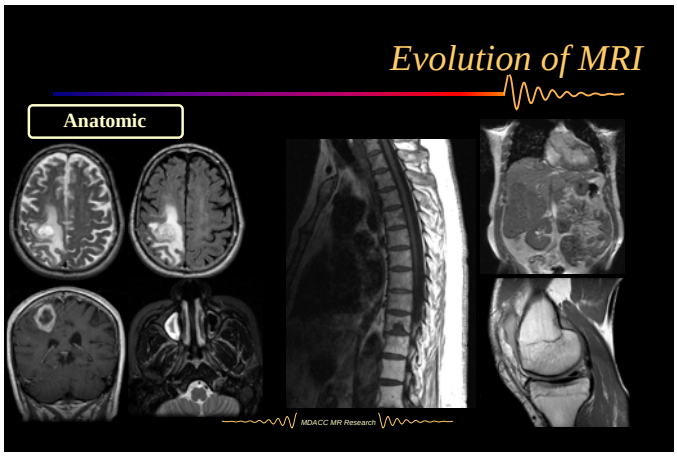
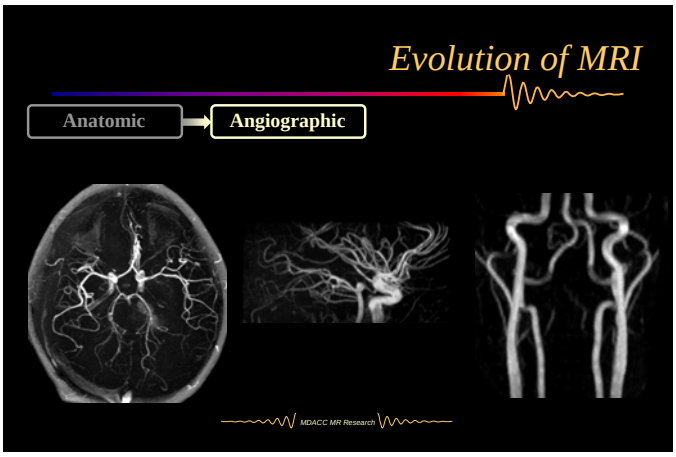
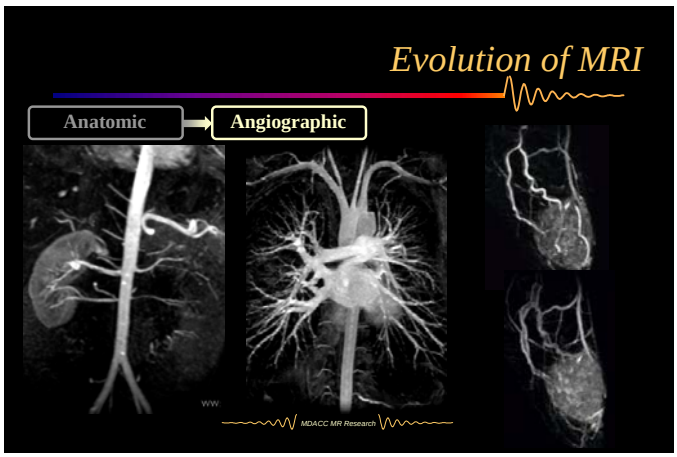
AC-PC Aligned View

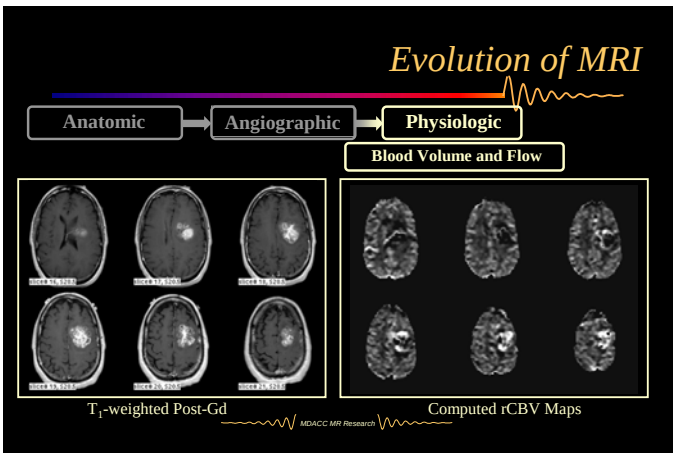

3D Display and Surface Vascular Mapping (3T Data)
Motor Mapping
Receptive Speech Mapping
MDACC MR Research


fMRI and surgical planning
http://www.stealthstation.com


Evolution of MRI
Anatomic
MDACC MR Research


Evolution of MRI
Anatomic
Angiographic
MDACC MR Research


Evolution of MRI
Anatomic
Angiographic
MDACC MR Research


Evolution of MRI
Anatomic
Angiographic
Physiologic
Blood Volume and Flow
T1-weighted Post-Gd
Computed rCBV Maps
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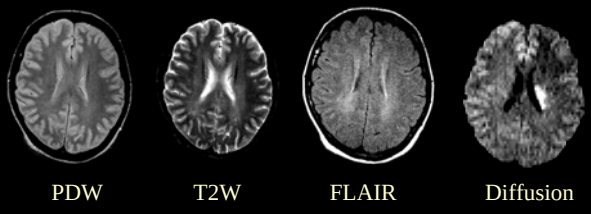
Evolution of MRI

Anatomic → Angiographic → **Physiologic**

Diffusion



PDW T2W FLAIR Diffusion

GE Medical Systems Applications Guide

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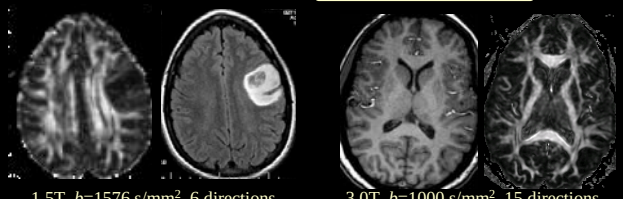
Evolution of MRI

Anatomic → Angiographic → **Physiologic**

Diffusion Tractography



1.5T, b=1576 s/mm^2, 6 directions 3.0T, b=1000 s/mm^2, 15 directions

MR Diffusion Tensor Imaging

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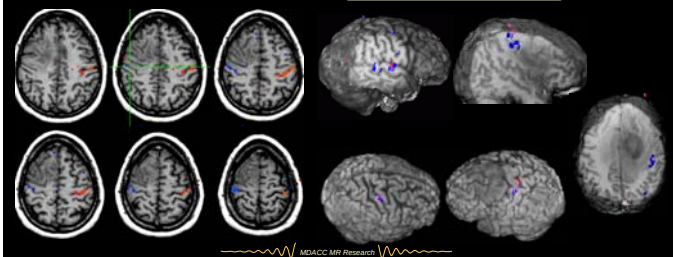
Evolution of MRI

Anatomic → Angiographic → **Physiologic**

Neuronal Activation



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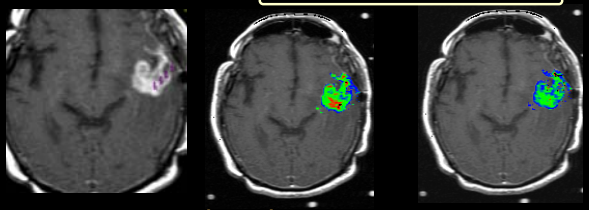
Evolution of MRI

Anatomic → Angiographic → **Physiologic**

Plasma Volume & Vascular Permeability



T1W Image Plasma Volume Fraction Map K^{trans} Map

Two-Compartment Pharmacokinetic Modeling of Contrast Uptake Kinetics

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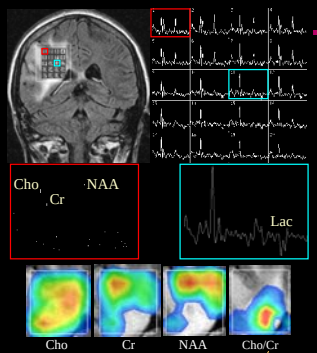
Evolution of MRI



^{1}H MR Spectroscopic Imaging Applications

NAA:	*N*-acetylaspartate
Cho:	Choline compounds
Cr:	Creatine/ phosphocreatine
Lac:	Lactate

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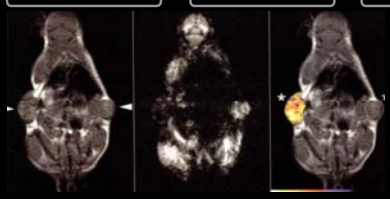
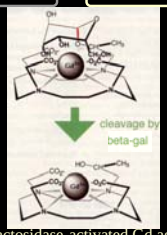
Evolution of MRI

Anatomic → Angiographic → Physiologic → **Molecular**



Transferrin receptor-targeted monocrystalline iron-oxide nanoparticles (MIONS)
Basilion *et al.* - MGH
Nature Medicine, March 2000



β-galactosidase-activated Gd agent
Meade *et al.* - Northwestern
Nature Biotechnology, March 2000

MDACC MR Research

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