Flow Effects in Magnetic Resonance (MR) Imaging, MR Angiography, MR Perfusion

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Overview of flow imaging

MR flow measurements can be distinguished in

- **Visualization of flow (Angiography, MRA)**
- **Quantitative flow measurement (q-flow)**

Measurement is done either for phase (phase contrast between moving tissue and stationary tissue), or for magnitude (magnitude contrast between moving and stationary tissue).

Q-flow, Phase Contrast

Principle of PC MRI: Use the influence of gradient fields on the signal phase to recover position, velocity, and acceleration of the flowing tissue (blood in arteries and veins of body and brain, CSF).

Relation between velocity and phase for the unipolar gradient

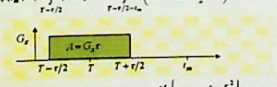
Position of blood $r(t) = r_0 + v_0(t - t_0) + \frac{1}{2}a(t - t_0)^2$ Taylor 2nd Order

Motion and gradient in x-direction with phase-measurement of the spins at position $r(t_m) = x(t_m) = x_m$ at time $t_m = t_0$

The phase at $t = t_m$ is given by (Substitute $t \mapsto t' = t - t_m$)

$$\varphi(t_m) = \gamma \int_{T-\tau/2}^{T+\tau/2} x(t) \cdot G_x dt = \int_{T-\tau/2-t_m}^{T+\tau/2-t_m} \left(x_m + v_m t' + \frac{1}{2} a t'^2\right) G_x dt'$$

For

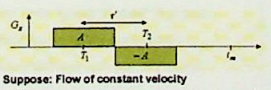


We get

$$\varphi(t_m) = \gamma x_m A + \gamma v_m A (T - t_m) + \gamma \frac{aA}{2}\left((T - t_m)^2 + \frac{\tau^2}{12}\right)$$

Source:Vlaardingerbroek et al.

Relation between velocity and phase for the bipolar gradient



Suppose: Flow of constant velocity

$$\varphi = \gamma A x(T_1) - \gamma A x(T_2) = \gamma v A (T_2 - T_1) = \gamma v A \tau'$$

With the *First Order Moment of the* bipolar gradient

$$A\tau' = m_1 = \int G(t) \cdot t \, dt$$

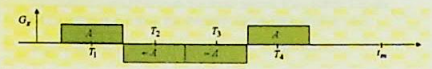
The areas of the bipolar gradient have to be same, so that the phase of spins in stationary tissue is not changed:

The *Zero Order Moment vanishes*

$$m_0 = \int G(t) \, dt = 0$$

Source:Vlaardingerbroek et al.

Relationship between velocity and phase for the back-to-back bipolar gradient



Two bipolar gradients with opposite First Order Moments lead to a phase independent on velocity and position

$$m_0 = m_1 = 0$$

The ***back-to-back bipolar gradients*** can be used for the design of velocity-insensitive sequences

In case of substraction of an image of a velocity sensitive sequence from an image from a velocity insensitive sequence, only the flow remains (PC-Angio)

Source:Vlaardingerbroek et al.

Accelerated flow and use of a bipolar gradient

Since velocity depends on time, only velocities at a certain point in time can be measured. Evaluation of the phase evolution for a bipolar gradient yields

$$\varphi(t_m) = \gamma \cdot v(t_v) \cdot A \cdot (T_1 - T_2) + \gamma \cdot a \cdot A \left[t_v \cdot (T_2 - T_1) + \frac{1}{2} \left(T_1^2 - T_2^2 \right) \right]$$

The time of expansion, where the phase gets independent on acceleration, is given by

$$t_v = \frac{1}{2}(T_1 + T_2)$$ Time, where the velocity is determined

There we have $\varphi(t_m) = \gamma \cdot v(t_v) \cdot A \cdot (T_1 - T_2)$

$v(t_v)$ can be found for a corresponding bipolar gradient this way

A bipolar gradient allows to relate the velocity in the center of gravity of the gradient to the phase of the spins

Source:Vlaardingerbroek et al.

Accelerated flow and the back-to-back bipolar gradient

For a simple back-to-back bipolarer gradient with 3 areas we have

$$m_0 = \int G(t) \cdot dt = 0$$

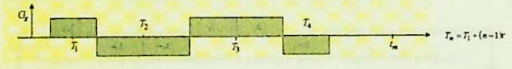
$$m_1 = \int G(t) \cdot t \, dt = 0$$

$$m_2 = \int G(t) \cdot t^2 \, dt = 4 \cdot A \cdot \tau^2$$

The relationship between phase and velocity given by

$$\varphi = \gamma \cdot a \cdot m_2$$

vanishes for multiple back-to-back bipolar gradients with 4 areas



Source:Vlaardingerbroek et al.

Field inhomogenieties and Eddy currents

$$\varphi(t) = \gamma \int \{ \alpha \mathcal{B} + \dot{\alpha} \mathcal{B}_0 + x \dot{G}_x(t) \} dt$$

B0 Inhom. — Eddy Current

Suitable gradient coil

Measurement 2x with different m_1

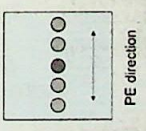
Source:Vlaardingerbroek et al.

Artifacts I: Signal extinction

Flow void: Signal extinction is provoked by strong velocity gradients. Strong velocity gradients apear beyond stenoses, bifurcations, or in regions with turbulent flow. The intra-voxel dephasing then leads to *„flow voids"*. This can be reduced or even avoided by the choice of short echo times.

Artifacts II: Ghosting

Ghosts: One finds "ghost images" in phase-encoding direction,since the periodically pulsating flow interferes with the MR-signal.



Artery perpendicular to the image

Source:Vlaardingerbroek et al.

Artifacts III: Misregistration

The position is encoded when the blood arrives at the center T of the phase encoding gradient. For the unipolar gradient, we then have:

$\varphi(t_m) = \gamma\, y(t_m)\, A_y + \gamma\, v_y\, A T - t_m)$ especially for $t_m = (n-1)TR + TE$

On the other hand, the spins get at position $y(T)$ a phase-shift by the encoding gradient of

$\varphi = \gamma\, n\, t_y\, G_y\, y(T) = \gamma\, n\, A\, y(T)$

It follows the relationship $y(T) = y(TE) + v_y (T - TE)$

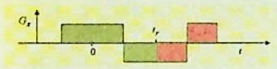
At the time of the echo, blood at $[x(TE), y(TE)]$ is observed actually at $[x(TE), y(T)] = [x(TE), y(TE) + v_y (T - TE)]$

Flow compensation

Subtraction method:

- Use a flow sensitive sequence to get an image
- Use a flow insensitive sequence to get an image
- Compute the difference between the images

I Selection gradient



Between t = 0 und t = t_r bipolar, thus flow sensitive:

$$\varphi(t_m) = \gamma\, v\, (t_r\, kt\, (T_1 - T_2) + \gamma\, a\, A \left[t_r (T_2 - T_1) + \frac{1}{2}\left(T_1^2 - T_2^2\right) \right]$$

Flow insensitivity can be established by adding a second, invers bipolar gradient.

Source:Vlaardingerbroek et al.

II Readout direction

At the time of the echo the zero order moment of the gradient vanishes, as well for SE as for GRE imaging

In case of a bipolar gradient, the phase of the moving spins becomes velocity dependent. To get independence, a second bipolar gradient can be added.

Source:Vlaardingerbroek et al.

III Phase encoding direction

In phase encoding direction flow results in spatial misregistration. Correcting this misregistration is possible while encoding the phase in a way as if it were measured at t=TE. For each phase encoding step an extra phase

$$\Delta\varphi = \gamma\, v_y A\, r' \quad \text{(bipolar Gradient)}$$

is to be added. This can be done by adding an extra-bipolar gradient. The corresponding first order moment then can be computed as

$$m_1 = A_B(n)\, r' = n\, A \cdot (TE - T)$$

and results in a position encoding that doesn't depend on velocity.

If there are no extra bipolar gradients added, the position encoding of the first echo of imaging sequences becomes velocity dependent.

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Part II: MRA (MR Angiography)

1. Phase contrast angiography (PCA)
2. Magnitude contrast angiography (MCA)
3. Artifacts
4. "Time of flight"-method

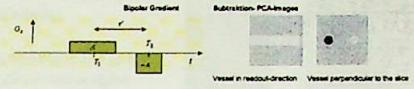
PC Angio: Subtraction method

Phase contrast methods are subtraction methods.

1. Take a flow insensitive image.
2. Take 3 flow sensitive images with flow sensitivity in all 3 directions.
3. Subtract the flow insensitive image from the flow sensitive ones.
4. Compute a magnitude image from the 3 images.

Source:Vlaardingerbroek et al.

Phase-contrast angiography. Quantitative Flow



Subtraction of flow insensitive GRE images from flow sensitive ones allows to determine the velocity of flow and to draw quantitativ flow maps.

The method is given by the velocity vector

$$I_{QCflow} = i\{\varphi(I_{xx}) - \varphi(I_1)\} + j\{\varphi(I_{yy}) - \varphi(I_1)\} + k\{\varphi(I_{zz}) - \varphi(I_1)\}$$

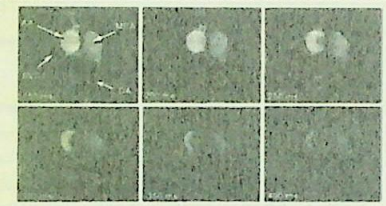
or, expressed by gradients and flow velocities

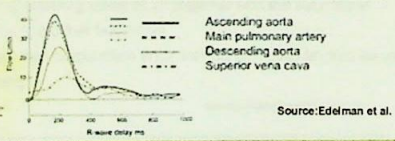
$$I_{QCflow} = iA_x c_x v_x + jA_y c_y v_y + kA_z c_z v_z$$

SE cannot be used dueto outflow between the excitation and the refocussing pulse.

Source:Vlaardingerbroek et al.

Phase-contrast angiography. Quantitative Flow





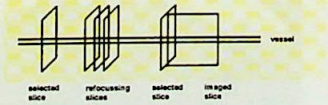
Source:Edelman et al.

Magnitude contrast angiography

I. 'Time-of-flight' or 'Inflow' Angiography
II. Contrast-Enhanced (CE) Angiography
III. Magnetisation Preparation
IV. 'Black-Blood' Angiography
V. Artifacts in MCA

I ‚TOF' or ‚Inflow' Angiography

Time-of-flight Method:
(a) Refocussing slices parallel to the selected slice.
(b) Refocussing slice parallel to the vessel.



Chose a refocussing rf-pulse in a refocussing slice. Using several refocusing slices at various positions, this allows to assess a velocity profile of the flow.

The velocity profile allows to compute the amount of blood flowing to the slice.

Source:Vlaardingerbroek et al.

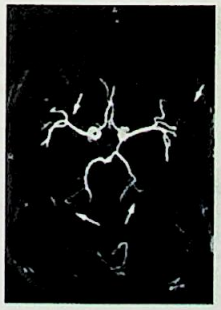
I ‚TOF' or ‚Inflow' Angiography

3 D Time-of-flight (TOF) image of the circle of Willis.

TR/TE/FA = 40/6/23°

Distal flow is better retained with smaller FA (see the arrows).

Acquisiton with higher FA (45°) leads to background suppression.

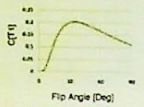


Source:Edelman et al.

II Contrast-Enhanced Angiography

CE-MRA Techniques use the shortening of T1 in blood by CA. T1-FFE spoiled gradient-recalled echo-Sequences areused. The transversal magnetization in steady-state (after a few HF-Pulses) is given by

$$M_T(x,y) = M_0(x,y)\frac{\sin\alpha\left(1-E_1\exp\left(\frac{-TE}{T_2^*}\right)\right)}{(1-E_1\cos\alpha)} \quad \text{with} \quad E_1 = \exp(-TR/T_1)$$



Steady-State Contrast C(T1) =(MT(blood)-MT(background))
Blood+CA (T1 = 48 ms; T2 = 28 ms)
Stationary tissue (T1 = 700 ms, T2 = 50 ms)
for GRE sequence parameters TR = 7 ms, TE= 4 ms

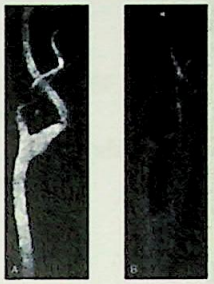
The time between injection and measurement is important!

II Contrast-Enhanced Angiography

Maximum Intensity Projection (MIP) of two consecutive acquisitions of the carotid arteries.

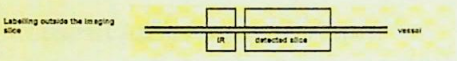
A: Appropriate timing with strong arterial and low venous signal.

B: Arterial and venous signal are of same intensity and difficult to differentiate.



Source:Edelman et al.

III Magnetisation Preparation



- Preparation pulses can be used to enhance the contrast between stationary and moving tissue
- Contrast for inflow-MRA can also be enhanced by MT (Magnetization Transfer) - Pulses
- Labeling inflowing blood by IR together with the subtraction technique is another technique
- The fact that oxygenation enhances T2 of blood can also be used for labeling

Source:Vlaardingerbroek et al.

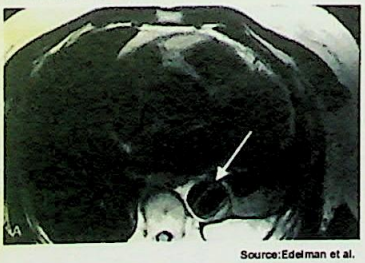
IV Black-Blood Angiography

Black blood: The SE technique can be used for the "black blood" signal extinction of flowing blood. The blood therefore has flown out of the imaging region at the time the second (inversion) pulse is applied. The remaining blood has not experienced the first (excitation) pulse and therefore does not give a signal. The signal thus disappears, and the flowing blood appears black.

The SNR remains somewhat limited with this technique.

IV Black-Blood Angiography

Black blood technique: The arrow shows an aortic dissection.



Source:Edelman et al.

V Artifacts in MC Angiography

Pulsating flow: Artifacts are present in the change of vessel diameter (10-15%) due to pulsating flow.

Contrast Agent (CA) inflow: Imaging during inflow of CA to the imaging region.

Turbulent flow: Turbulent flow leads to signal extinction.

Magnetic Resonance Perfusion and Diffusion Imaging, functional MRI

Bernd Müller-Bierl, PhD, Dipl. Phys.
CRAD – MR Center

Perfusion Sensitive Sequences

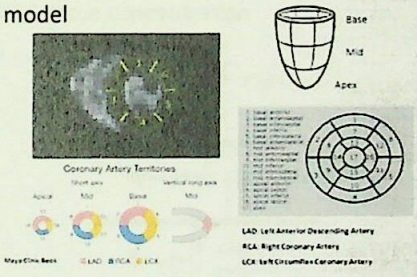
- In MRI: Measurement of AIF, TRF using CA, T1w MR Seq. with SI ∝ CA
- First Pass, Dual Bolus, Fast/Ultrafast Imaging, SSFP and EPI, Ex. Turbo FLASH

Tracers

- Contrast Agent (CA), Transit Times
- In the past: Radioactive Tracers as O^{17}, H^2 (Deut) (Endogenous)
- Exogenous like Gd-DTPA (Magnevist), Molecular Weight
- Without CA App., Ex. ASL

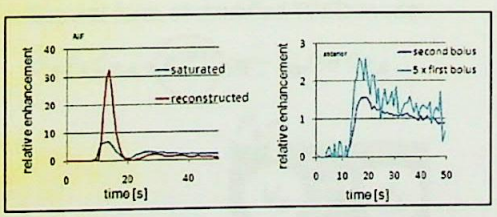
Example outside Brain

The 17 segment heart model

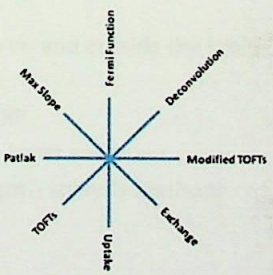


Data acquisition and analysis

AIF and TRF



The 8 perfusion flow analysis methods



Functional Analysis

- Functional or Operator
- Lives in Sobolev or Hilbert Spaces
- Example: The antiderivative

$$F(x) = \int_a^x f(\xi)\, d\xi$$

$$F' = f$$

Convolution

- Definition

$$F(k) \otimes G(k) = \frac{1}{\sqrt{2\pi}} \int_{-\infty}^{+\infty} F(k-k')\, G(k')\, dk'$$

'k-space'

- Then it holds that

$$f(x) \cdot g(x) \leftrightarrow F(k) \otimes G(k)$$

Used for solving DEs

Algorithm

Examples of algorithms giving a number:

- Babylonian Algorithm (1700 BC)
- Newton-Raphson
- Levenberg-Marquardt

Stopping problem: An algorithm always terminates!

Fermi Function

- Total tissue concentration

$$C = F_P R \otimes C_{P,A}$$

Fermi-Dirac Distribution Function

$$R(t) = \{\exp((t - t_0 - t_d)k + 1)\}^{-1} u(t - t_d)$$

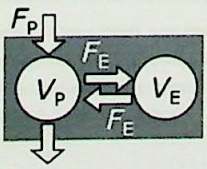
Heaviside or Integrated Dirac Function

$$u(t - t_d) = \begin{cases} 0 & t < t_d \\ 1 & t > t_d \end{cases}$$

Modified TOFTs

- General form of modified TOFTs model:

$$C = V_P C_{P,A} + K^{trans} \exp(-k_{ep} t) \otimes C_{P,A}$$

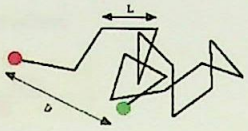


Resume

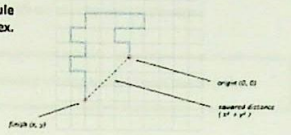
- Perfusion in- and outside the brain
- Tracers
- AIF and TRF
- Examples heart and brain
- The 8 (eight!) analysis methods

Random Walk

Brownian motion



Unlimited motion of one molecule in an ensemble of molecules (ex. AIR)



Brownian motion

The Einstein Diffusion Equation describes microscopic transport of particles and heat:

$$\partial_t\, p(r,t|r_0,t_0) \;=\; \frac{\sigma^2}{2\gamma^2}\,\nabla^2\, p(r,t|r_0,t_0)$$

Brownian motion

From the EDE, we obtain as spatial distribution of particles in ex. AIR:

$$\left\langle (r(t)-r(t_0))^2 \right\rangle \;=\; 6\frac{\sigma^2}{2\gamma^2}\,t$$

$$D=\frac{\sigma^2}{2\gamma^2}$$

$$\left\langle r^2 \right\rangle = 2\,Dt$$

$$\left\langle r^2 \right\rangle = 4\,Dt$$

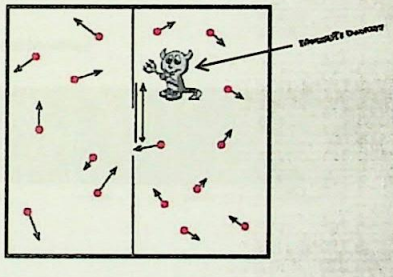
$$\left\langle r^2 \right\rangle = 6\,Dt$$

During diffusion, entropy increases.

Entropie

Entropie is a measure of disorder

Linked terms:
Energie
Probability
Information



Cybernetics

... society can only be understood through a study of the messages and communication facilities which belong to it; and that in the future development of these messages and communication facilities, **messages between man and machines, between machines and man, and between machine and machine,** are destined to play an ever increasing part. (p.15)

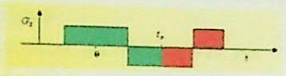
from the article "**Cybernetics in History**", Wiener (**1954**), cited after Maggie Mc Garry, "Norbert Wiener's Cybernetic Theory and Parental Control", URL http://www.colorado.edu/communication/meta-discourses/Papers/App_Papers/McGarry.htm, **Feb 14 2013**.

Mathematics for flow imaging

- Flow independent sequences
- Flow compensation
- Subtraction method
- Phase or Magnitude

Mathematics for flow imaging

Example: Slice selection gradient



Between t = 0 und t = t, bipolar, thus flow sensitive:

The signal phase depends on velocity and acceleration.

Flow insensitivity can be established by adding a second, invers bipolar gradient.

Mathematics for diffusion

The decay of the Transverse Magnetization (TM)

Bloch Equation for TM:

$$\frac{dM_T}{dt} = -j\gamma \vec{r} \vec{G} M_T - \frac{M_T}{T_2}$$

Transport Process:

$$\vec{j}_{M_T} = \vec{D}\nabla M_T$$

Mathematics for diffusion

Solution for the Transverse Magnetization (TM)

$$M_T(t) = M_T(0)\exp\left[-\int_0^T \left(\vec{k}^T(t')\vec{D}\vec{k}(t')dt'\right)\right]\exp\left(-j\vec{k}(t)\vec{r} - \frac{t}{T_2}\right)$$

In case 1D

In this case we can write the TM as

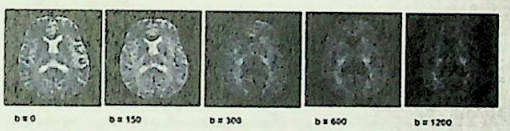
$$M_T(TE) = M_T(0)\exp(-b_x D_{xx})\exp\left(-\frac{TE}{T_2}\right)$$

with

$$b_x = \int_0^{TE} k_x^2(t)dt = \gamma^2\int_0^{TE}\left[\int_0^t G_x(t')dt'\right]^2 dt$$

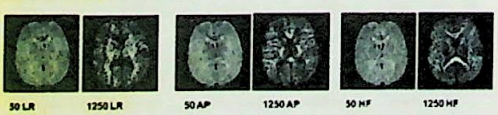
Measurement of diffusion

Images for different b-values



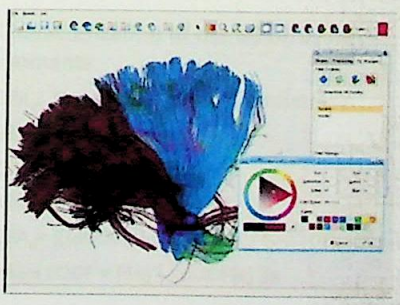
Measurement of diffusion

Diffusion weighted images



D_x Trace

Fiber Tracking in White Matter



Multiple Fibers

Diffusion measures *dispersion of water molecules* within a few tens of milliseconds described by a probability function: **p**

White matter axon radii: **0.1 - 10 µm**

voxel: **1 - 5 mm**

Heuristic rule **b = ADC^{-1}**

Fiber orientation distribution function: **fODF** (describes the fraction of fibers in orientation)

Diffusion orientation distribution function: **dODF** (describes probability of diffusion in orientation)

Multiple Fibers

Approach (model-based): Describe the dispersion by a weighted sumof gaussians:

volume fraction, Displacement, Diffusion time, Gaussian with covariance $2\,\mathbf{D}_i\,t$

$$p(x) = \sum_{i=1}^{n} a_i \, G(\mathbf{x}, \mathbf{D}_i, t)$$

The normalized diffusion weighted signal then is given by

wavevector in direction of grad B

$$A(\mathbf{q}) = \sum_{i=1}^{n} a_i \exp\left\{-t\,\mathbf{q}^T \mathbf{D}_i\,\mathbf{q}\right\}$$

E.g. n = 2: 2 x 6 +1 (a_1=1-a_2) parameters

Multiple Fibers

volume fraction, Displacement, Diffusion time, Gaussian with covariance $2\,\mathbf{D}_i\,t$

$$p(x) = \sum_{i=1}^{n} a_i \, G(\mathbf{x}, \mathbf{D}_i, t)$$

wavevector in direction of grad B

$$A(\mathbf{q}) = \sum_{i=1}^{n} a_i \exp\left\{-t\,\mathbf{q}^T \mathbf{D}_i\,\mathbf{q}\right\}$$

Solution by additional constraints on the diffusion:

(a) Enforce positive definiteness

(b) Enforce cylindrical symmetrie

(c) Fix the DT's Eigenvalues

(d) Ensure voxel-to-voxel coherence by spatial regularization

(e) Isotropic/ One fiber/ two fiber automatic model selection

Multiple Fibers

Non parametric approaches (model-free): Distinguish fanning or bending fibers from straight ones.

Diffusion Spectrum Imaging: For infinitely short pulses we have p = FT{A}.

Q-Ball Imaging: Interpolate the dODF found by using a linear basis of spherical functions.

Spherical Deconvolution: Measure A (Diffusion weighted Signal) and R (Convolution of the measurement). The fiber orientation can be obtained by deconvolving A (**q**) = $\int$ f(**x**) R(**q**, **x**) d**x**.

Persisten Angular Structure: Find the angular structure by a maximum entropy principle. PAS is deconvolution with R(**q**, **x**) = r^{2}cos(r**q**.**x**).

Diffusion Orientation Transform: The DOT is a single contour of p at fixed radius R0. It calculates a variant of the dODF.

Multiple Fibers

Method	Acquisition requirement	Computation time	Accuracy	Bias
Two tensor	low/medium	medium	medium	low
Ball and stick	low	medium	medium	low
PAS MRI	medium	hight	high	low
SD (low-pass)	medium	low/ medium	medium	medium
SD (cSD)	medium	medium	medium	low
DSI	very hight	medium	high	medium
Q-Ball	medium/ hight	low/ medium	medium/ low	medium
DOT	medium/ hight	medium	medium	medium

Multiple Fibers

Characterization of the **dODF** by its moments:

GFA_2: Generalized fractional anisotropy

GFA_3: Generalized skewness

GFA_4: Generalized kurtosis

with moment n being given by

$GFA_n = \{((dODF(\mathbf{x}) - DODF)^n \, d\mathbf{x}) / (\int (dODF(x))^n \, d\mathbf{x})\}^{1/n}$

where $DODF = (4\pi)^{-1} \int dODF(\mathbf{x}) \, d\mathbf{x}$

Fiber Tractography in White Matter

1. Brain function relies on white matter pathways.
2. Diffusion is preferred along fiber orientation in axon nerve-fibers.
3. White matter pathways can be found in the living brain by Diffusion Tractography.

Fiber Tractography in White Matter

Tractography Algorithms for the reconstruction of diffusion in white matter pathways:

Local <> Global

Deterministic <> Probabilistic

Model Based <> Model Free

Simple <> Complex

Fiber Tractography in White Matter

Example: **Streamline Tractography**

A streamline is a 3D curve in space with the tangent being the first eigenvector of the diffusion tensor.

$$\frac{d\mathbf{r}}{ds} = \varepsilon_1(\mathbf{r}(s))$$ Differential eqn.

The solution is based on interpolation (Runge-Kutta).

Fiber Tractography in White Matter

Errors in **Streamline Tractography** can be classified in

1. Imaging noise due to poor estimation of diffusion directions
2. Modeling errors by chosing the wrong model for the voxel
3. (Numerical) integration errors

Fiber Tractography in White Matter

The stopping problem in **Streamline Tractography** can be solved by

1. Minimum FA allowed
2. Maximum curvature allowed

Probabilistic Tractography can be used to avoid stopping in regions with high uncertainty:

(model, data) → Compute the x% confidence that the path of least hindrance to diffusion is from A to B.

Introduce an **uncertainty Orientation Distribution Function**: The **uODF** can be computed using various methods as bootstrapping, Bayesian methods, functional approximation and calibration.

Fiber Tractography in White Matter

Choice of local description:

- Use many different gradient orientation (~100 000) and a complex description for the local model.

Designing a tractography study:

- Reconstruct many path ways
- Relate functional information and related white matter pathways

Future advances:

- Global tractography
- Use the whole fODF
- Combine tractography with quantitative models

Resume

- Random Walk and Brownian Motion
- Entropy and Cybernetics
- Flow Imaging
- Diffusion Imaging
- White Brain Matter

fMRI of the brain

Bernd Müller-Bierl, PhD, Dipl. Phys.

Historical approach

1936 Magnetic state of hemoglobin changes with its state of oxygenation (Pauling and Coryell)

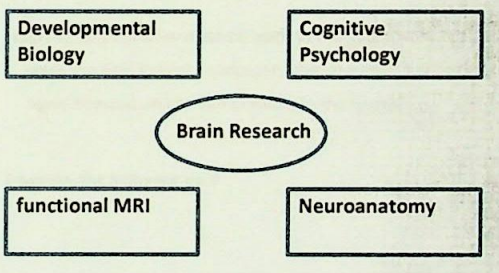
1945 Nuclear magnetic resonance effect by Purcell, Torrey and Pound, and, independently, by Bloch

1982 Transverse relaxation rate (T2) changes in blood depending on the oxygenation (Thuiborn)

1990 Signal enhancement in rat brain during blood oxygen level increase (Ogawa)

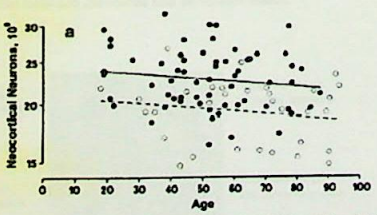
1992 fMRI signal changes during visual stimulation (Menon, Ogawa, Kim, Ellermann, Merkle, Tank, Ugurbil)

Brain Research



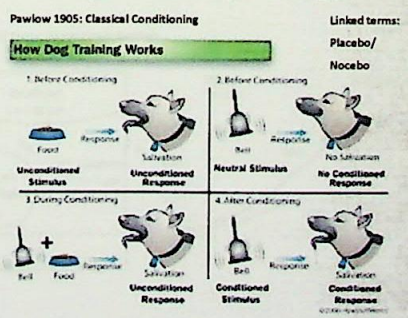
Source: Muller-Bierl (2013)

Developmental Biology



Number of neurons in human brain, depending on age (years). Black circles are men, white circles are woman (Pakkenberg & Gundersen, THE JOURNAL OF COMPARATIVE NEUROLOGY, 1997).

Cognitive Psychology



Neuroanatomy

Human brain – white matter and grey matter

Source:Wikipedia

fMRI: The BOLD effect

Neural activity in brain

- ⇒ local cerebral blood flow (CBF) increases more than the cerebral metabolic rate of oxygen (CMRO2)
- ⇒ oxygen extraction fraction (E) decreases with activation of the brain
- ⇒ ⇒ local blood is more oxygenated
- ⇒ ⇒ there is less deoxyhaemoglobin present in nearby tissue
- ⇒ ⇒ the local MR signal increases

Susceptibility effect for GRE and SE

Susceptibility Artifacts in Spin-Echo and Gradient-Echo Imaging

STEFAN POSSE AND WALTER P. AUE

NMR Unit of the Medical Faculty, University of Berne, Bühlrainstr 28, 3012 Berne, Switzerland

Received August 9, 1989; revised January 4, 1990

A general model for the quantitation of spin-echo and gradient-echo artifacts in samples with macroscopic susceptibility inhomogeneities is developed. Its usefulness is tested by comparing experimental images and computer simulations of cylindrical susceptibility inhomogeneities. Positional displacements, gradient-echo phase shifts, and diffusion and sampling time effects, which are responsible for local image distortions and signal losses, are calculated on a subpixel basis to provide a quantitative comparison of susceptibility artifacts under various imaging conditions. The formulation will be useful for the interpretation of susceptibility-related signal losses in gradient-echo imaging and for the analysis of spin-echo susceptibility artifacts at high field strengths. In addition susceptibility effects in NMR microscopy and contrast properties of particulate contrast agents may be simulated. © 1990 Academic Press, Inc.

Susceptibility effect

Magnetic susceptibility material leads to misregistration, this leads to signal intensity distortion, that expresses itself as signal attenuation and compression in the image.

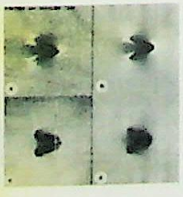
Example: the following slide

Susceptibility effect for GRE and SE

Comparision between measured and computed effect.

Measured image.
SE based sequence.

Measured image.
GRE based sequence.



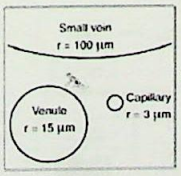
Computed image.
SE based sequence.

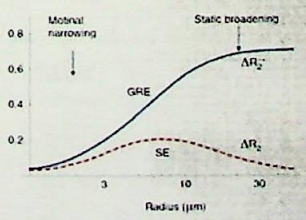
Computed image.
GRE based sequence.

Posse and Aue 1990

Vessels in the brain ('Vein effect'):

Technical term: **Blood Brain Barrier**

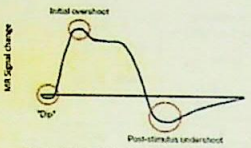




Source: Edelman et al. (Eds.)

BOLD-model ('Brain effect')

Balloon model (Buxton 1998): The venous compartment is treated as a distensible balloon.



A typical BOLD response for a block design stimulus. Transients are an occasional brief initial dip at the beginning, an initial overshoot, and a prolonged post-stimulus undershoot.

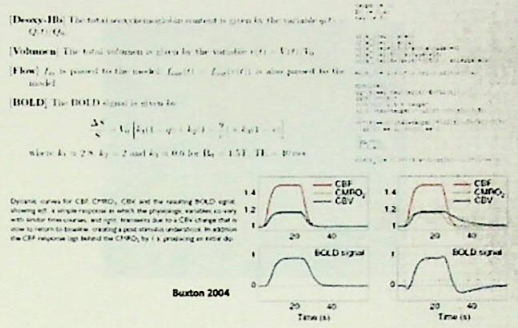
$$\frac{dq}{dt} = \frac{1}{\tau_{MTT}}\left[f(t)\frac{E(t)}{E_0} - \frac{q(t)}{v(t)^{\alpha-1}} f_{out}(v,t)\right]$$

$$\frac{dv}{dt} = \frac{1}{\tau_{MTT}}\left[f(t) - f_{out}(v,t)\right]$$

$$E(f) = 1 - (1-E_0)^{1/f}$$

Source: Edelman et al. (Eds.)

BOLD-model



Buxton 2004

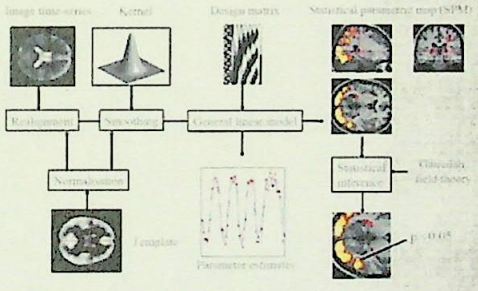
Activation maps

We need to do substraction measurement on stimulated and non-stimulated brain.

How to compute activation maps from the substraction measurement?

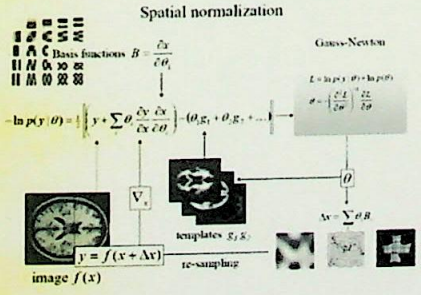
By statistical evaluation of the data!

Data analysis



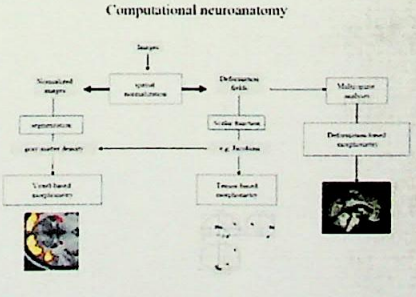
Friston, Experimental Design and Statistical Parametric Mapping

Spatial Normalization



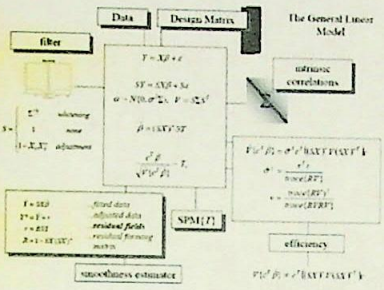
Friston, Experimental Design and Statistical Parametric Mapping

Computational Neuroanatomy



Friston, Experimental Design and Statistical Parametric Mapping

General Linear Model



Friston, Experimental Design and Statistical Parametric Mapping

Thank you very much!