1 Introduction

Since 1977 when the first clinical MRI scanner was patented nuclear magnetic resonance imaging is increasingly being used for medical diagnosis and in scientific research and application in practice. In 1992 Ogawa and Turner showed image contrast in changing oxygenation state of blood and opened a unique method of investigation. Functional MRI is a technique to localize brain areas that are activated during a specific task. Using this methodology, it is possible to construct whole brain activation maps for sensory and mental functions with high spatial resolution. The most important role of fMRI in investigating human brain function arises from the fact that brain function is spatially segmented. This functional specialization can be defined and mapped by fMRI utilizing secondary hemodynamic an metabolic responses to alterations in neuronal activity. An important additional feature of fMRI is capability to follow signal changes in real time, even through temporal as well as spatial resolution of fMRI is dictated by the characteristics of the hemodynamic response. Due to fMRI we can get the whole of brain image in several hundred msec with spatial resolution of received image is about a few mm. Although there is a Brain Spectroscopy, which gives opportunity to see brain activity through metabolites, its spatial resolution is only sq.sm, in some cases it isn't good enough.

2 The basic principles of NMR

At the heart of the techniques is magnetic property of some nuclei. One of those kinds of nuclei is proton. Owing to interaction of proton magnetic moments with external magnetic field macroscopic magnetization is appearing in sample.μ_i magnetic moment of nucleus N_i, N_0 - quantity of nuclei in sample (Figure 1). Macroscopic magnetization (M_0) aligns along external field. If M_0 is disturbed from equilibrium frequency of its nutation Is $\omega = \gamma B_0$ (Larmor frequency) Summary magnetization aligns along field direction there is not magnetization in transversal plane. RF pulse has Larmor frequency and it turns M_0 . If whole M_0 is appearing in orthogonal plane that means it was a 90^0 RF pulse. Two time constant T_1 and T_2, are used to describe the evolution of a nuclear spin system after being disturbed from its equilibrium (a vector M_0 along B_0) T_1 is known as the spin-lattice (longitudinal) relaxation time. It describes the return of the disturbed magnetization to its thermal equilibrium along B_0. T_2 is called the spin-spin (or transverse) relaxation time because it describes the decay in phase coherence between the individual spins in the transverse plane, resulting signal loss. Actually dephasing of transverse magnetization is due to both: microscopic molecular interactions (this exponential decay is described by T_2) and spatial variations of the external field ΔB_0 T_2^* is characteristic time in such conditions. T_2^* is about hundred msec in 1-1.5 T and it is shorter for higher B_0. Signal depends from value of transversal magnetization or, other words, from sum of magnetizations from all volumes of this sample (Figure 1). Local inhomogeneities of external magnetic fields speed up decay of signal.

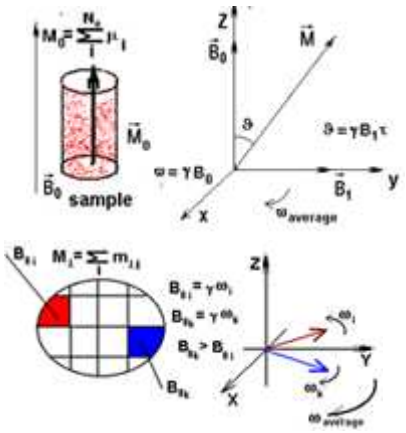

Figure 1: Magnetization and dephasing of transversal magnetizations

3 Image Creation

In MRI image of 3D subject is receiving slice by slice. For selection layer is using gradient. Gradient is applying magnetic field which value has a linear dependence on coordinate(Figure 2). It acts during RF impulse is on in orthogonal direction of selected slice. Due to it RF impulse excites only spins which Larmor frequency coincides with diapason of pulses frequencies. The width of a layer depends on value of gradient and frequency diapason of pulse. For getting 2D information in selected plane is using a method to encode space coordinate. The spatial position of a spin is encoded by applying gradient along chosen axis. For example axis x. On frequency axis each frequency corresponds to certain coordinate (Figure 2). For the second dimension phase - encoding gradient is

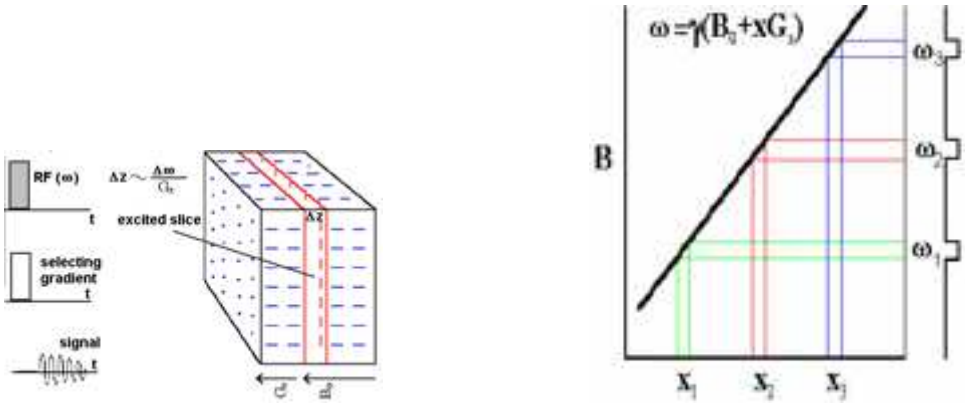

Figure 2: Slice selection and frequency encoding

applying. The gradient duration is changing and it changes signal. Such way is getting 2D matrix of data. Every line there is receiving by measurement signal FID in present of Frequency-encoding gradient in real time and each line

correspond to different duration or amplitude of it. After 2 D Fourier transformation we've got a massive on frequency plane where every frequency variable is proportional to coordinate corresponding to direction of applying gradient. Graphic view of this massive is spatial distribution of initial data. Usually in practice amplitude of Frequency-encoding gradient changes amplitude but not its duration. Sequence for image receiving is in Figure 3. For more comfortable accepting information was invented a k-space matrix (Figure 4).

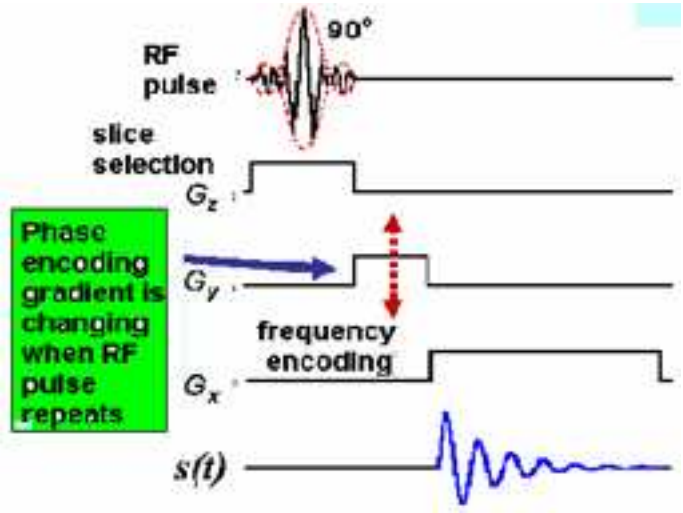

Figure 3: Sequence for receiving image

4 Gradient Echo

In the gradient present FID is too fast. For more qualitative signal is applying a gradient echo signal. You already know mechanism of echo, when dephased magnetizations collected by 180^0 pulse. In Gradient echo after gradient dephased transversal magnetizations, opposite sign gradient collects them back (Figure 5). Sequence for gradient echo signal is shown on Figure 5. The amplitude of this signal depends on inhomogeneities of constant field which is not coursed by gradients, what means, that signal depends from T_2^*. One echo signal is filling one line of k-space. It means that for matrix 128x64 we need 64 echo signals in present of 64 different value phase gradient. In medical investiganion factor of experiment time is very important. Especially it's important for fMRI where people is watching for process dynamic. Echo Planar Imaging (EPI) sequence was developed to faster a receiving data. Due it is possible to get a plane image during one RF pulse acting. Frequency-encoding gradient changes its sign

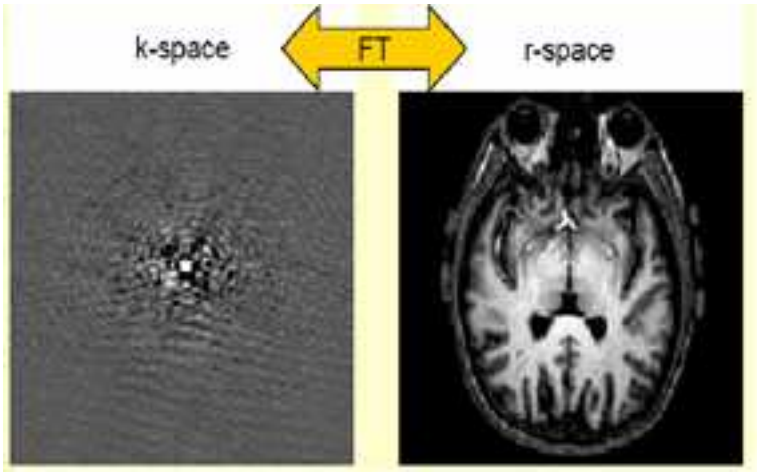

Figure 4: K-space and R-space

several times and phase-encoding gradient repeats every changing such way is getting a distinguish in signal phase during every reading act. During one pulse there is an opportunity to get several echo signals each one in the present of own value phase gradient. Figure 6 shows a sequence for receiving a slice image by using EPI.

5 Necessary equipment

Figures 7-9 demonstrate device for MRI.

6 Blood Oxygenation Level Dependent signal (BOLD)

BOLD is a special method founded on contrasting image by T_2^* BOLD stands for blood oxygenation level dependent. The mechanism of the BOLD contrast is the following. By giving up oxygen, the hemoglobin iron in blood undergoes a change in spin state from diamagnetic low spin in the oxygenated state to paramagnetic high spin in the deoxygenated state. (Molecule of hemoglobin is on Figure 10) With increases in arterial well-oxygenated blood flow the venous blood pool shows a significant decrease in paramagnetic deoxyhemoglobin concentration. This decrease in paramagnetic leads in turn to less intravoxel dephasing, and as result to signal increases with T_2^*weighted sequences. In a brains active zones a signal increase is only 21.5T) due to the BOLD effect. The effect is proportional to the field strength. That is why for fMRI is using fields 1.5 T and hirer. In strong fields inhomogeneities is more significant. Different

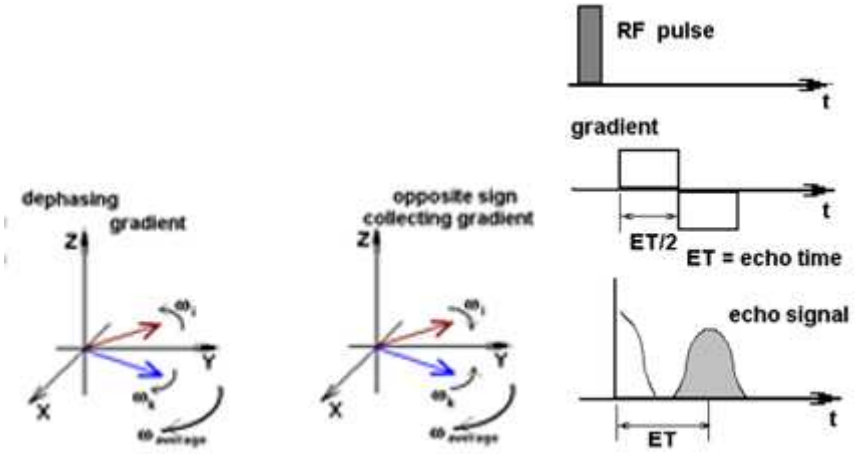

Figure 5: Mechanism gradient echo and sequence for it

between signals amplitudes can be seen by comparing graphic acquired during rest to graphic acquired during stimulation (Figure 12). Selection of optimum echo time is important for acquisition a reliable data. In fMRI, a large number of images - tens to several hundreds - are measured consecutively in a single experiment lasting anywhere from a few minutes to a quarter of an hour. The collected data are a time series of signal intensity from small volume elements or "voxels" covering regions of interest or the whole brain. During the data acquisition period, inputs for brain activation a presented to the subject in the magnet at appropriate period. The input can be sensory stimulation, sensory input - guided cognitive tasks, subject - initiated mental activity, or even spontaneous brain activity the subject may not be aware of. Image signals responding to the input are then compared with the control image signal. It is preferable to image the whole brain simultaneously so that all the relevant activation patterns can be captured at once. Therefore, ultrafast imaging techniques, which may require extra MRI hardware, are indispensable. With echoplanar imaging the magnetization induced from a slice in the brain by a single excitation can be measured at 64x64 2-D complex data point in as short as 30 msec to produce one image. Within a few seconds, the whole brain area can be covered and the same image acquisition sequence is repeated until the desired time series data are collected. For getting more precisely information we must choose a time when mental activity gives maximum signal (Figure 13). Researching shows that it's a 4-6 sec after simulation started. There are some investigations of human brain in high magnetic fields (4T) High fields have advantages and disadvantages in fMRI. By increasing the static magnetic field from 1.5 to 4 Tesla or higher, MR signals generally increase because the magnetization of sample is proportional to the field strength. The increase of MR signals improves the signal to noise ratio. In addition, it's expected that signal source is more localized at higher fields because the BOLD effect from the veins and venules is proportional to the

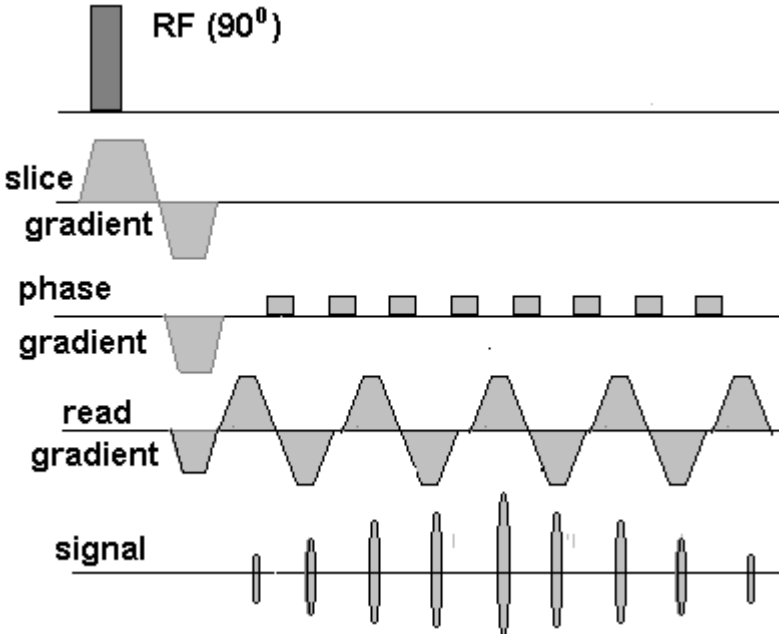

Figure 6: Echo Planar Imaging (EPI) sequence

field strength. Based on these two factors is expected to have higher spatial resolution at higher fields. There are a disadvantages of fMRI at higher fields. The decay time of magnetization signal due to dephasing (T_2^*) is shorter at higher fields, witch leads to more limited time for measuring signals after single frequency pulse. It is thus more difficult to implement the fast imaging techniques, such as echo planar imaging. Furthermore, the susceptibility artifacts due to the inhomogeneity of the statistic magnetic fields around nasal cavity and ear cavity are lager at higher fields. Considerable signals are lost there, especially when the fast imaging techniques are employed.

7 Some examples of applying fMRI)

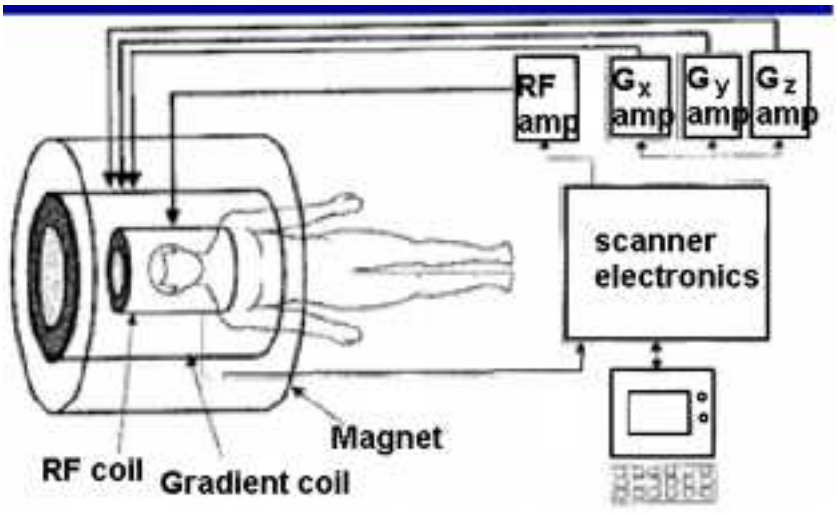

Figure 7: Scheme of Equipment

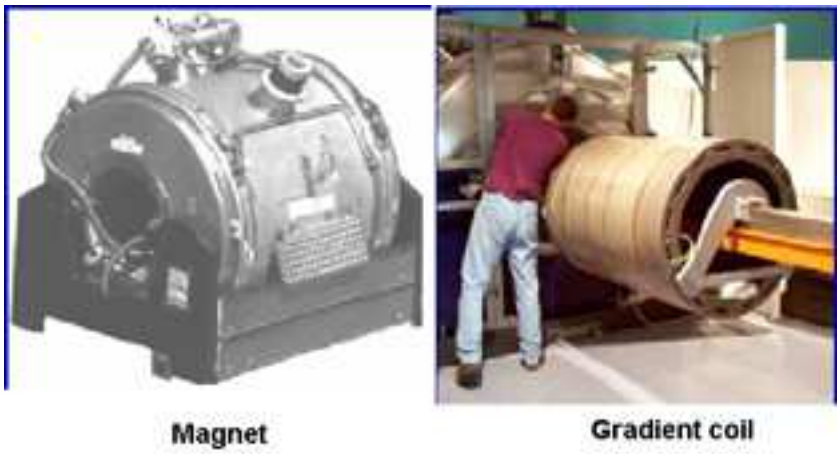

Figure 8: magnet and gradient coil photos

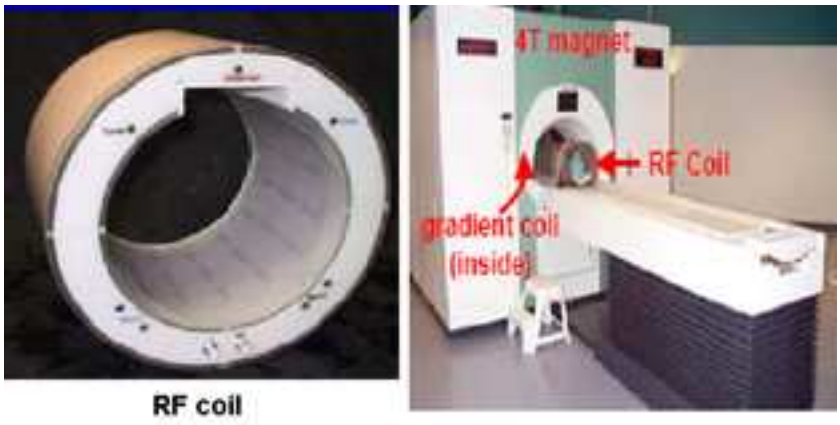

Figure 9: RF coil and MRI device photos

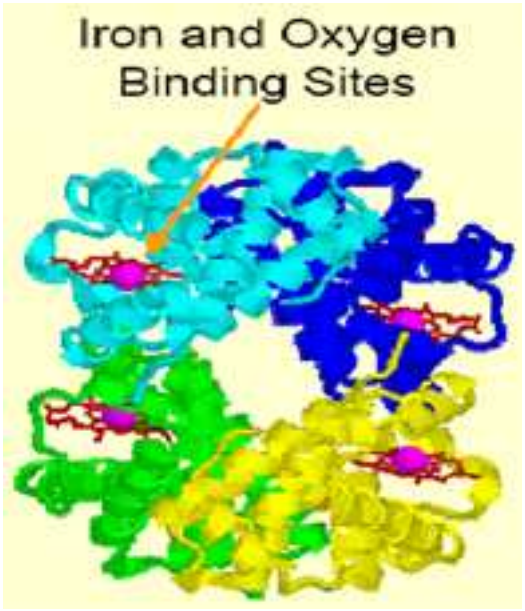

Figure 10: hemoglobin molecule

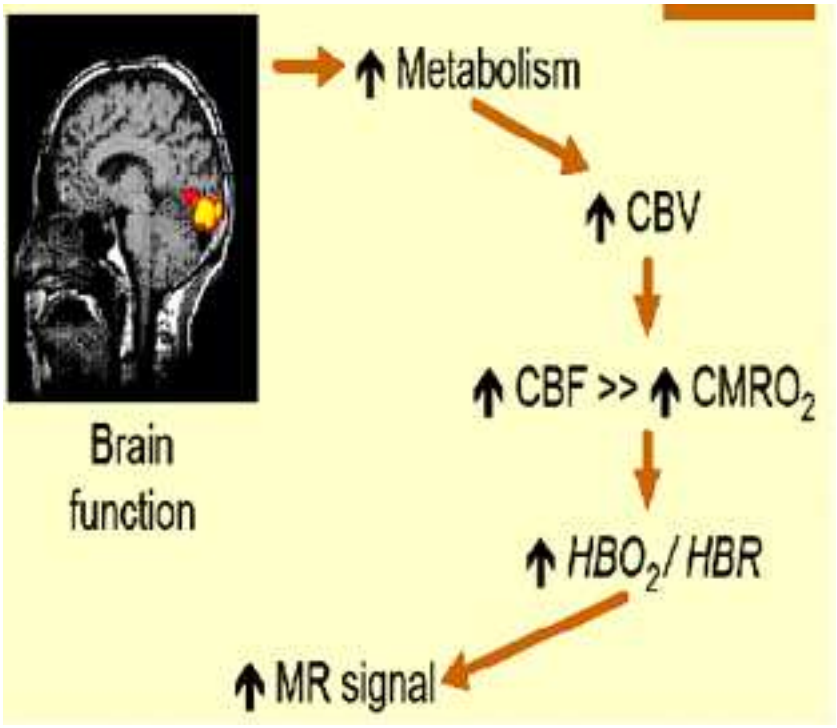

Figure 11: Mechanism BOLD

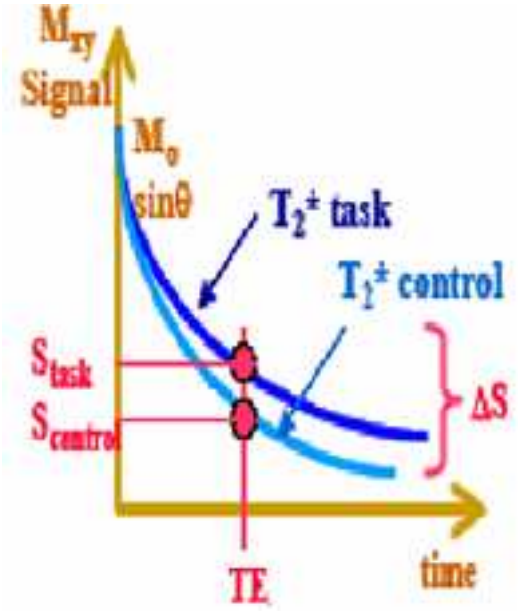

Figure 12: Difference between control and task signal amplitudes

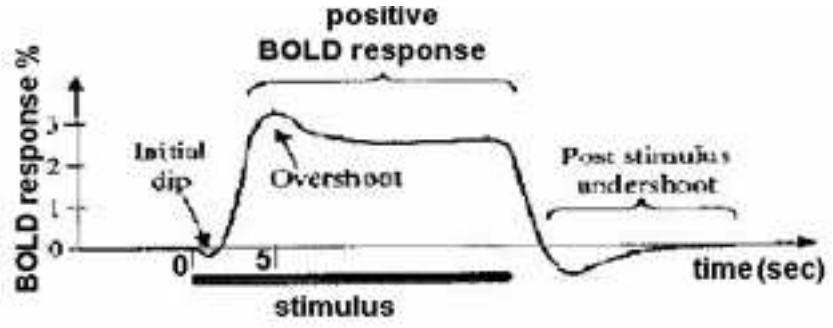

Figure 13: Hemodynamic response function

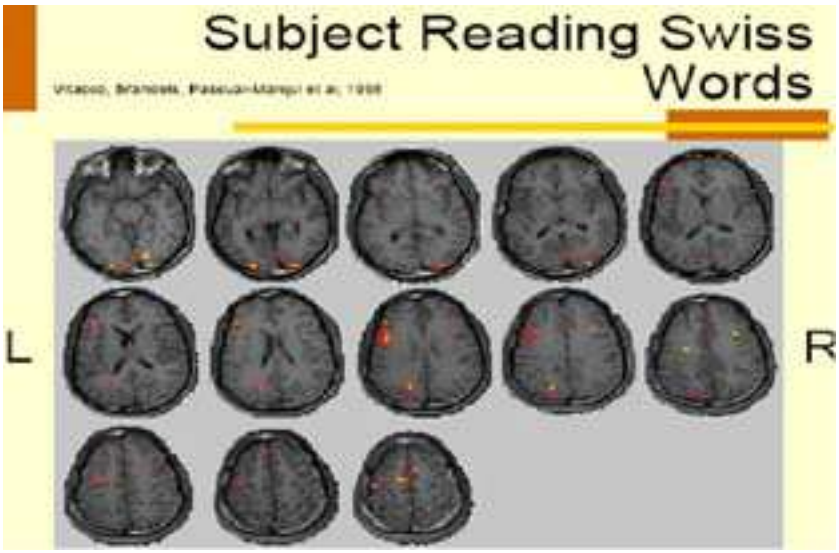

Figure 14: Active zones of whole brain slice by slice. Subject adult and healthy
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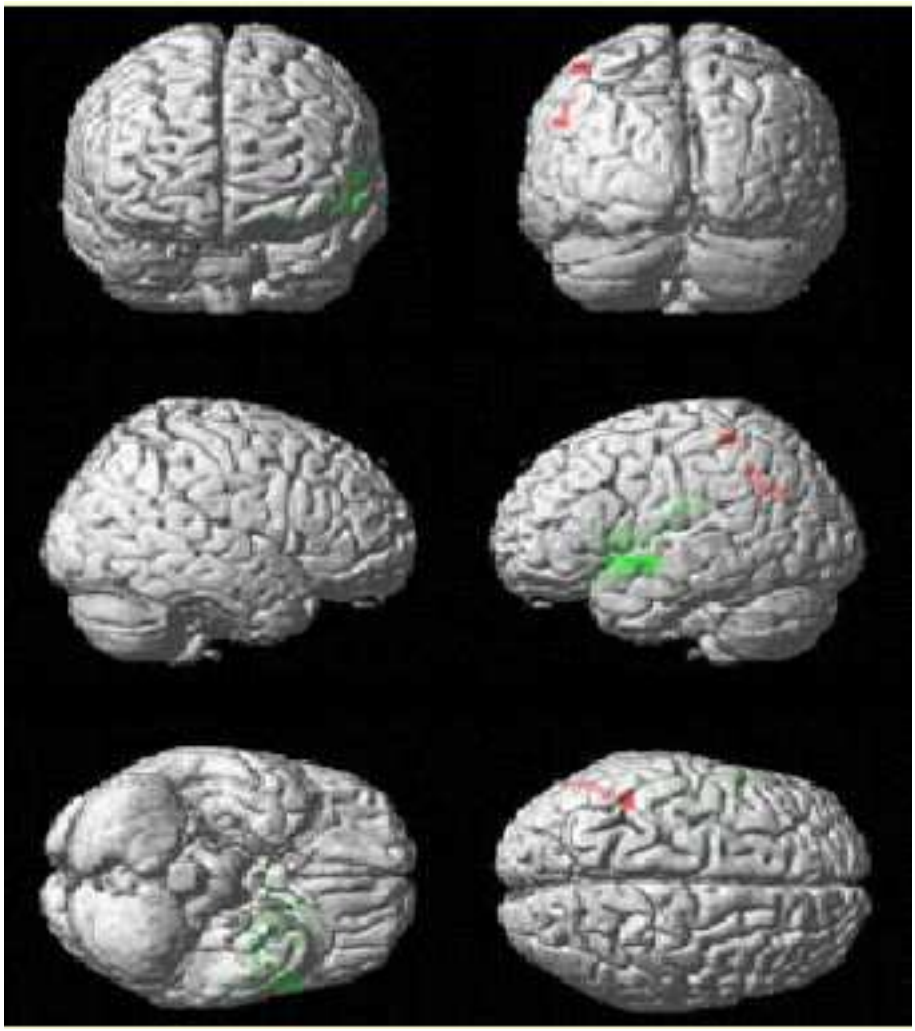

Figure 15: Calculation in adults and children. Red label is adults, green label is a normal achieving school children.

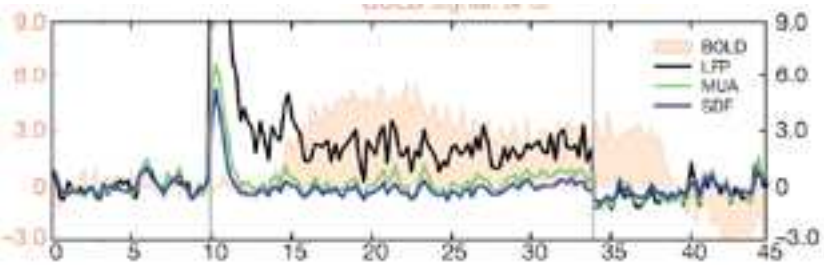

Figure 16: Correlation between BOLD and LFP (local field potential).