2 MR Imaging

2.1 Introduction and History

This chapter provides a brief introduction into magnetic resonance imaging as a medical imaging modality and as the basic technique used for the acquisition of the images for which the concepts developed throughout this thesis are developed. The purpose of this chapter is twofold. One the one hand the reader should be given a brief introduction into the basic concepts and modern developments of magnetic resonance imaging (MRI or NMR[1]). On the other hand, from a point of view of a reader with a background in image processing, it is important to be aware of the possibilities and limitations of the MR image acquisition process. Often people with profound knowledge in image processing make an enormous effort to come up with intricate solutions for processing and analysis of MR images. However, tuning the acquisition side, e.g. by making changes in the pulse sequence, the scanning parameters resp. protocol or the image reconstruction from the raw data, the problems could be solved easily without sophisticated post-processing algorithms. For example, the segmentation of a particular anatomical structure may be impossible given a standard image contrast. By making use of another MR imaging contrast, the segmentation can be done by a simple thresholding or windowing of the grey values. It is therefore important in MR image processing to keep an eye on both the acquisition and the processing of the images, the more so in the case of this thesis where steps of human interaction the examination workflow should be automatized by means of image processing algorithms.

Magnetic resonance imaging is a modality for in-vivo imaging of structures or functions of the body. The term magnetic arises from its basic principle of the interaction of nuclear magnetic spins with external magnetic fields. The external magnetic fields may be static or varying over space and time. In the latter case they are also referred to as magnetic gradients resp. as a sequence of gradients. The term resonance insinuates the interaction with external radio frequency fields. In simple words, MR data represent the induction signal of macroscopic magnetization of aligned nuclear spins in receiver coils.

The huge success of MRI is based on two facts: First, the 'non-invasive' nature of magnetic fields allows for examinations without harming effects and without the use of ionizing radiation. Second, MRI is extremely rich in information, as its data may not only deliver information about the anatomical structure but also about physical properties like flow, temperature,

[1] nuclear magnetic resonance

diffusion, etc. Physiological effects, like stimulation of brain activity as in fMRI [HBTV99] can be observed, or even metabolic processes of certain parts of tissue, as in MR spectroscopy applications.

Historically, the roots of NMR go back to the 1940s. Bloch [BHP46] and Purcell[PTP46] described the quantum mechanical effects of nuclear spins in magnetic fields in 1946. They managed to measure the signals of water and paraffin and were awarded the Nobel Prize for their contributions in 1952. In 1973, Lauterbur [Lau73] and Mansfield [MK73] laid the foundation for MR as an imaging technique by making spatial encoding possible, which 30 years later resulted in being awarded the Nobel Prize for their breakthrough concepts.

The value of MRI as a diagnostic medical imaging modality was first discovered by Damadian [Dam71]. He pointed out the value of MRI for medicine by examining that tumors can have significant effects on the behavior of the magnetic signal, which means that lesions and tumors may be detected from irregularities in the MR images.

2.2 Comparison to Computed Tomography

In common discussions, there is often a lot of confusion about the differences and similarities of magnetic resonance imaging (MRI) and computed tomography (CT). Often the terms are mixed up, and even if there is knowledge about the physical principles of the image acquisition that the two modalities are based on, people are not familiar with the particularities of the two. Often, it is assumed that the two modalities could produce equal images or that they could be used interchangeably. As a matter of fact, this is not the case. This section will explain in short terms the major particularities and differences between the two modalities.

The basic differences between MR and CT are best explained, if one considers the physical principles that the image acquisition is based on. In MR, the interactions of the nuclear spins with external magnetic fields and RF pulses are measured by detecting the resonance signal of nuclei after excitation. CT images are reconstructed from multiple projection images, acquired by a rotating X-Ray tube. Contrast in CT images is therefore mainly determined by the different densities of the tissues with respect to the absorption of the X-rays. As for MR images, contrast is determined by a whole set of parameters, like the density of protons in the tissue, their relaxation times, chemical properties, etc. Depending on the scanning technique, it is also possible to measure other physical values which may be related to physiological properties, like flow or temperature, diffusion in diffusion weighted imaging, brain activity in functional MRI, concentration of metabolites in chemical shift imaging, etc. [HBTV99].

In general, a CT exam is faster, cheaper and easier to perform than an MR exam. It is usually the first imaging modality of choice for emergency patients with multiple traumata. Because of its speed and high reproducibility it is also often and commonly used in many clinical standard examination procedures. There is no general answer to the question which modality is superior or produces better images. Both modalities have weaknesses and strengths. Basically, MR is better for contrasts in soft tissue [DHL+89], whereas CT produces better morphological contrasts. Therefore, MR is often superior for detection and identification of tumors while CT is unrivaled in detecting bones and calcification.

Both modalities have certain hazards for patients being examined. Patients with metallic

implants, cardiac pace makers or tattoos containing iron should not undergo MR examinations, because movement of these objects caused by the magnetic fields or induction heating may occur. Other hazards include claustrophobia from which patients being examined may suffer, eventual peripheral nerve stimulation in the case of magnetic gradients being switched too quickly and hyperthermia from absorption of radio frequency energy. Hyperthermia is avoided by consideration of the specific absorption rate (SAR), which defines the ratio of temporal increase in temperature caused by RF energy [DWRB02]. These calculations often require complex calculations and modeling. Many of the other effects are avoided by hardware and software limitations for the scanning parameters. Apart from that, no major negative long-term effects have been observed for MR examinations in contrast to CT where ionizing radiation is used, although there is a possibility of minimizing this effect by low-dose CT scans. As a consequence of the fact that no long-term harming effects have been observed for MR, there are no regulatory objections against acquiring additional MR images, which are of no diagnostic value. This allows for setting up image processing applications which require additional special images. Since these images are not further used, their contrast can be designed and optimized for the need of the application or processing algorithm. For instance, this is the case for the localizer images used in this thesis.

2.3 Basic Principles

2.3.1 Spins, Excitation and Relaxation

Spin Angular Moment

Depending on their composition of neutrons and protons, atom nuclei can have an intrinsic spin angular moment or shorter a spin. In accordance with quantum theory, this spin can only have discrete quantized values.

It turns out that nuclei with an impair number of protons and an impair number of neutrons have integral spin values (e.g. deuterium ^2H) and nuclei with with an even pair of protons and an impair number of neutrons have half-integral spin values (like the most abundant isotope of hydrogen ^1H [2]). Nuclei of atoms with even numbers of both protons and neutrons do not have any value of spin at all. The latter type of atoms is not suitable for MRI.

The quantized angular spin moment can be expressed by the following equation:

$$L = \hbar\sqrt{I(I+1)} \tag{2.1}$$

where L is the angular spin moment, $\hbar$ Planck's constant[3] divided by 2π and I the respective integral or half-integral quantum number.

The nuclear spins cause a magnetic field of a strength B that is directly proportional to the angular spin moment. The proportionality constant γ is called the gyromagnetic ratio, which is a specific constant of every isotope.

[2] since ^1H hydrogen nucleus consists of only one proton, it is common to use the expression proton equivalently

[3] $\hbar = 1.0545Js$

$$\vec{B} = \gamma \vec{L} \tag{2.2}$$

where $\vec{B}$ is the magnetic field strength, γ the gyromagnetic ratio and $\vec{L}$ the angular spin moment.

Because of its abundance in the human body and its high gyromagnetic ratio, ${}^1H$ is the most interesting isotope in terms of magnetic resonance imaging with a half-integral angular spin moment of $\frac{1}{2}$.

Another phenomenon of quantum mechanics is that the expectation values of the angular spin moments are not only quantized in their values but also in their orientations with respect to a specific axis. It turns out that, if the quantum number of the angular spin moment is I, there exist $2I + 1$ possibilities for the angular spin to orientate along a specific axis. These orientations are characterized by the $2I + 1$ values between $-I$ and $+I$. These values are also called the orientation quantum numbers.

For protons, for example, with an angular spin moment of $\frac{1}{2}$ there exist two orientations along a certain axis with the orientation quantum numbers $-\frac{1}{2}$ and $\frac{1}{2}$. For deuterium with an angular spin moment of 1 there exist three orientations with the orientation quantum numbers -1, 0 and $+1$, etc.

Interaction with an External Magnetic Field

Under normal conditions, the different states of orientation are energetically equivalent. Given the presence of an external magnetic field, however, an effect called Zeeman interaction can be observed: the energetic equivalence is split up into $2I + 1$ different states, one for each orientation quantum number along the axis defined by the direction of the given external magnetic field, as in figure 2.1 a): on the left side there is energetic equivalence, on the right side, under the presence of the external magnetic field B_0 the equivalence is split up in a state of anti-parallel orientation and high energy and a state of parallel orientation and low energy. The direction of the external magnetic field B_0 defines the z-axis of the coordinate system.

In a classical description, under the presence of an external magnetic field, the angular magnetic moment rotates or precesses around the axis of the magnetic field. The axis of precession is given by the direction of the z-axis as shown in figure 2.1. This phenomenon is caused by the physical law of the constancy of an angular moment which is the same that makes a spinner precess around the direction of gravity. The frequency of the precession is called the Larmor frequency:

$$\omega_L = \frac{\gamma B_0}{2\pi} \tag{2.3}$$

The difference in energy between the two states of spins of the protons is directly proportional to the Larmor frequency and therefore proportional to the external magnetic field as well:

$$\Delta E = h\omega_L = \hbar\gamma B_0 \tag{2.4}$$

where h is Planck's constant, $\hbar = \frac{h}{2\pi}$ as before, B_0 the magnetic field strength and ω_L the respective Larmor frequency.

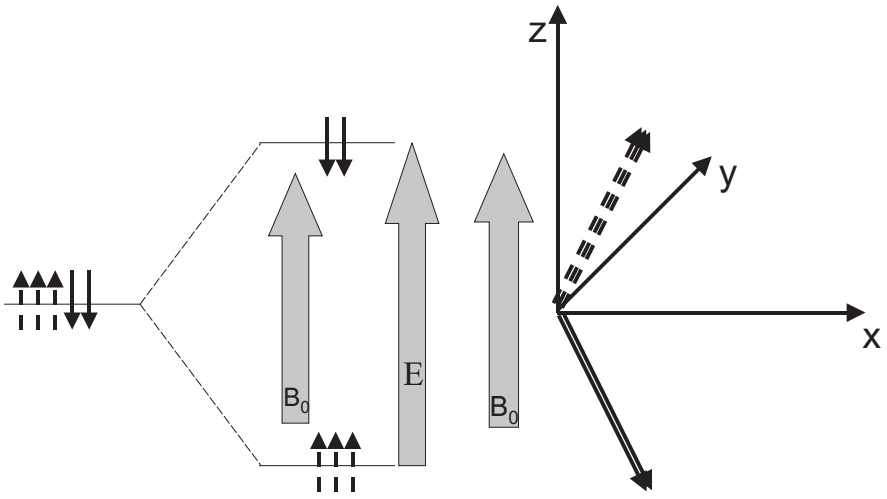

Figure 2.1: a) shows a Zeeman diagram. Whereas on the left side there is energetic equivalence of the up and down spin states of a proton. On the right, situation under the presence of an external magnetic field B_0. The state with spin parallel to the external magnetic field is energetically lower than the state anti-parallel. b) shows the orientation of the spins parallel (dashed) or anti-parallel (solid) to the external magnetic field B_0. The direction of B_0 determines the z-axis of the given coordinate frame.

In MR experiments the induction signal caused by precessing protons is measured. The vast majority of the protons in biological tissues are bound either in water or in fat. Due to their different chemical surroundings their Larmor frequencies differ by a small value. This difference is called the chemical shift and can be a source for artifacts as section 2.5.6 explains.

Macroscopic Description and MR Signal

Because of the energy difference between the two states, they are not occupied equally by the protons. The occupation difference can be quantized by a result from statistical thermodynamics in the so called Boltzmann equation:

$$\frac{N_{upper}}{N_{lower}} = e^{-\frac{\Delta E}{kT}} \tag{2.5}$$

In this equation N_{upper} and N_{lower} describe the number of nuclei in the upper respectively lower level of energy. ΔE describes the energy difference between the states. T is the temperature and k the Boltzmann constant[4].

At room temperature of 293 K and a magnetic field strength B of 1 Tesla there is an excess of $1 : 10^6$ nuclei in the lower state in comparison to the upper state. To give an idea, in a probe of 9 g water containing roughly $6 \cdot 10^{23}$ protons, this means $6 \cdot 10^{17}$ protons. This is enough to

[4] $k = 1.3806 \frac{J}{K}$

observe a macroscopic net magnetization of the tissue under influence of the external magnetic field: since more nuclei have their spins oriented parallel to external magnetic field than in the opposite direction, the tissue has experienced a net magnetization in the direction of the external magnetic field. That is the macroscopic description which is the overall vectorial sum of all nuclear magnetic moments. The stronger the basic magnetic field, the larger the excess of magnetization. Most modern clinical MR systems have magnetic fields of 1.5 or 3 Tesla. However, there are also research systems with higher basic magnetic fields.

As the spins of the nuclei precess incoherently, their phases sum up to 0, so there is no macroscopic net magnetization in a plane perpendicular to the basic magnetic field. As the B_0 direction is the z axis, this perpendicular plane is called the transversal plane or the xy-plane. If all nuclei were in phase, a net magnetization in the xy-plane precessing around the z-axis could be observed. The next section shows how phase coherence and net magnetization in the xy-plane can be achieved by means of RF pulses.

The measured signal in an MR experiment stems from the detection of the electromotive force induced by the precessing magnetization. As the receiver coils are aligned perpendicular to the xy-plane, the transversal net magnetization M_{xy} is the one which is measured as a complex MR signal with the x component being the real and the y component being the imaginary part.

RF Pulses

An RF pulse or simply pulse can be described as an electro-magnetic wave with periodic magnetic content B_1, interfering with the net magnetization M and causing a rotation of the net magnetization. The angle of the rotation of the magnetization is determined by the length, the strength and the bandwidth of the pulse. The rotation axis is determined by the directions of B_0 and B_1.

For example, a 90° pulse causes the net magnetization in z-direction to rotate into the xy-plane, as delineated in figure 2.2. In the same way, other pulses, for example 180° pulses can be applied, which cause the z magnetization to be inverted into -z direction.

After the application of the pulse the protons will gradually return to the equilibrium state, which is a net magnetization in z direction, in the same way that a ball positioned at a steep hill rolls downhill towards the lowest state of energy. This phenomenon is called relaxation and has various reasons, among them the tendency for energy stability. The process of applying a pulse to achieve a higher energetic state by rotating the net magnetization is called excitation.

Spin Lattice Relaxation

Spin lattice relaxation or longitudinal relaxation is the process in which macroscopic net magnetization along the z axis is restored. It is an enthalpy driven process, and it can be described by an exponential decay:

$$M_z(t) = M_{z_0} \cdot e^{-\frac{t}{T_1}} \tag{2.6}$$

$M_z(t)$ is the net magnetization in z direction after time t. M_{z_0} is the net magnetization in z direction before the experiment starts at $t = 0$. T_1 is defined as the time needed by the system

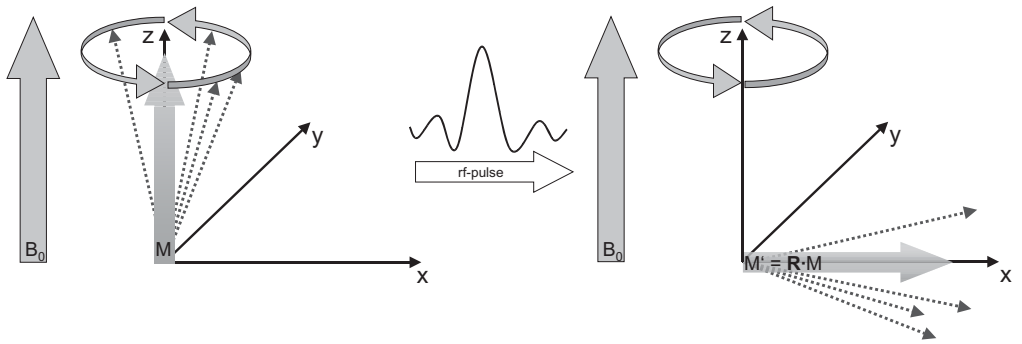

Figure 2.2: The effect of a 90° RF-pulse on net magnetization M. The direction of M is rotated according to the rotation matrix R perpendicular to B_0 and the rotation axis. R is defined by the length and magnitude of the pulse. In this case it is a 90° rotation. The rotated net magnetization M' keeps precessing around the z-axis.

to recover all but $1/e$ of the original net magnetization in z direction.

Considering T_1 relaxation is very important to understand the effects of subsequent pulse application experiments. If the delay time between two pulses is smaller than $3T_1$, which normally is the case, it is insufficient for the tissue to recover all its initial z magnetization. Therefore following pulses will result in considerably less net absorption and less measurable signal. T_1 depends on the type of tissue and its composition of fat and water but also on certain extrinsic factors. T_1 can be measured using the inversion recovery experiment [HBTV99]. Typical values of T_1 for biological tissues are typically in the range of seconds.

Spin Spin Relaxation

Spin spin relaxation or transversal relaxation is an entropy driven process in contrast to longitudinal relaxation. Entropy is a measure of disorder of a system and according to the second fundamental law of thermodynamics, entropy in a closed system increases steadily towards a maximum value. After the application of the pulse and the rotation of net magnetization into the xy-plane all precessing protons are in phase, which corresponds to a high state of order. As time elapses and the protons precess according to their Larmor frequencies this coherence of phases gradually gets lost as the state of high order decreases. Eventually, all protons will return to be out of phase. This state causes zero net magnetization because the vectors in the transversal xy-plane sum up to 0.

$$M_{xy}(t) = M_{xy0} \cdot e^{-\frac{t}{T_2}} \tag{2.7}$$

where M_{xy} is the net magnetization in the xy-plane immediately after the application of the pulse and T_2 the time after which the net magnetization has decreased to $1/e$ of its initial value.

The value of T_2 also depends on the type of tissue and additional extrinsic factors. Typically, liquids have small values of T_2, while solids exhibit long T_2 times. This makes liquid structures appear bright in T_2 weighted images.