ARTICLE IN PRESS

Journal of Magnetic Resonance xxx (2018) xxx–xxx

Contents lists available at ScienceDirect

Journal of Magnetic Resonance

journal homepage: www.elsevier.com/locate/jmr

RF pulse methods for use with surface coils: Frequency-modulated pulses and parallel transmission

Michael Garwood *, Kamil Uğurbil

Center for Magnetic Resonance Research and Department of Radiology, University of Minnesota, Minneapolis, MN 55455 USA

ARTICLE INFO

Article history:
Received 2 January 2018
Accepted 24 January 2018
Available online xxxx

Keywords:
MRI
MRS
Surface coil
B1 inhomogeneity
Array coil
Radiofrequency pulse
Frequency-modulated
Adiabatic pulse
Parallel transmission
pTx
Ultrahigh field

ABSTRACT

The first use of a surface coil to obtain a ^{31}P NMR spectrum from an intact rat by Ackerman and colleagues initiated a revolution in magnetic resonance imaging (MRI) and spectroscopy (MRS). Today, we take it for granted that one can detect signals in regions external to an RF coil; at the time, however, this concept was most unusual. In the approximately four decade long period since its introduction, this simple idea gave birth to an increasing number of innovations that has led to transformative changes in the way we collect data in an *in vivo* magnetic resonance experiment, particularly with MRI of humans. These innovations include spatial localization and/or encoding based on the non-uniform B_1 field generated by the surface coil, leading to new spectroscopic localization methods, image acceleration, and unique RF pulses that deal with B_1 inhomogeneities and even reduce power deposition. Without the surface coil, many of the major technological advances that define the extraordinary success of MRI in clinical diagnosis and in biomedical research, as exemplified by projects like the Human Connectome Project, would not have been possible.

1. Introduction

The first *in vivo* surface coil ^{31}P NMR spectra published by Ackerman et al. [1] initiated a revolution in magnetic resonance imaging (MRI) and spectroscopy (MRS). Today, we take it for granted that one can detect signals in regions external to an RF coil, but at the time this concept was most unusual since in every application ranging from chemistry in small samples to imaging humans, the conventional thinking required circumscribing volume coils where the sample was inserted into and contained within the coil. This unusual strategy of detecting signals adjacent to a simple loop of wire certainly ignited the field of *in vivo* MRS immediately. Magnetic resonance (MR) in general, but particularly MRS, suffers from limited sensitivity, a problem that is exceptionally pronounced when the volume of interest is actually much smaller than the coil and sample dimensions, as encountered when working with large samples contained within circumscribing RF coils. With its simple planar loop geometry, the surface coil provided for the first time a convenient new probe for *in vivo* MRS that yields high sensitivity for the tissue region immediately adjacent to it [2,3]. In addition, because its RF field (B_1) drops off in both radial and axial directions, early *in vivo* MRS studies exploited this spatial dependence of B_1 for achieving spatial localization without using pulsed B_0-gradient coils, which in those days were unshielded and produced large eddy currents that degraded spectral quality.

In the early days of surface coil MRS, much effort was devoted to developing B_1-localization techniques that could provide control in positioning the volume of interest precisely. NMR journals were soon publishing many articles introducing new localization techniques that exploited the spatially dependent B_1 of one or more surface coils. The main techniques of that period were called depth pulse [4,5], rotating frame zeugmatography (RFZ) [6–8], and Fourier series windows [9–11]. Still today, there are significant advantages to using B_1- versus B_0-gradient techniques. Unlike B_0-gradient fields, B_1 fields can be switched on and off rapidly (in 1–2 microseconds), and negligible eddy currents and acoustic noise are produced when pulsing B_1 coils. Furthermore, B_1-localization techniques can be performed without using a spin- or stimulated-echo acquisition, which inevitably forfeits signal from many low-gyromagnetic ratio (γ) nuclei that have spin-spin relaxation time (T_2) values comparable to or less than the minimum achievable echo time, TE.

Following the development of actively-shielded B_0-gradient coils, B_1-localization techniques were largely superseded by B_0-gradient methods due mainly to their greater flexibility in

* Corresponding author.
E-mail address: gar@umn.edu (M. Garwood).

https://doi.org/10.1016/j.jmr.2018.01.012
1090-7807/

Please cite this article in press as: M. Garwood, K. Uğurbil, RF pulse methods for use with surface coils: Frequency-modulated pulses and parallel transmission, J. Magn. Reson. (2018), https://doi.org/10.1016/j.jmr.2018.01.012

controlling the dimensions and placement of the MRS voxel. In response to these technological developments, our research focus turned to creating methods to produce uniform flip angles with RF coils that have highly inhomogeneous B_1, like a surface coil, using so-called adiabatic pulses [12–16]. Developments in adiabatic pulses, as well as in frequency-modulated pulse techniques in general, continue to take place at a rapid pace, and the number of applications where they play key roles is ever expanding for both MRS and MRI research.

In the past decade, interesting new approaches for performing MRI with B_1 gradients have been introduced based on the localized and spatially varying fields generated by arrays of surface coils. These techniques are based on parallel imaging (PI) strategies that are now commonly used to accelerate image acquisition in MRI (e.g., SENSE [17] and GRAPPA [18]). Due to recent developments in parallel transmit (pTx) capabilities (e.g., [19,20], and reviewed in [21]) with multi-coil arrays that can apply independent RF waveforms to each channel simultaneously, the prospects for achieving much greater localization precision and the potential for new effective B_1-gradient techniques has led to exciting possibilities in this area.

Herein we provide our perspectives on two important areas in biomedical MRI and MRS that can be linked directly to the introduction of surface coils by Ackerman and colleagues. Specifically, these are (1) frequency-modulated (FM) pulses, and (2) parallel transmission (pTx).

2. Overview of frequency-modulated pulses

Nowadays the types FM pulses and their applications are quite diverse. Some FM pulses, such as the popular chirp [22] and hyperbolic secant (HS) pulses [23], can function in an adiabatic manner, whereby the flip angle remains close to π above a user-defined threshold RF amplitude. Other FM pulses, like the 2D versions of chirp and HS pulses [24–26], function adiabatically only when taking certain trajectories in k-space. Of note, the first volume-selective (3D) adiabatic pulse, which was introduced only recently, can be implemented with pTx to enable undersampling of its 3D k-space trajectory and this can be used to shortened its overall duration [27].

Adiabatic π-pulses are commonly implemented in localized MRS sequences due to their ability to produce a uniform π flip angle over a broad bandwidth, without using a high level of peak RF power. One such sequence is known as LASER (localization by adiabatic selective refocusing) [16]. LASER and versions thereof (e.g., semi-LASER [28,29]) are commonly used nowadays to perform single-voxel MRS and spectroscopic imaging.

In MRI, FM pulses are playing key roles in novel spatiotemporal-encoding (SPEN) techniques [26,30,31] that offer the possibility to overcome many of the problems plaguing the widely used single-shot imaging method, EPI (echo planar imaging). The SPEN techniques exploit the fact that the phase of the transverse magnetization (M_{xy}) following a frequency-swept pulse is a quadratic function of the resonant frequency. By applying a B_0 gradient that imparts a spatially dependent resonant frequency, the vertex of this quadratic phase function can be controlled and moved at will through space as a function of time. SPEN techniques have been developed utilizing one or more adiabatic π-pulses, a non-adiabatic $\pi/2$-FM-pulse, or 2D versions of these. In 2D SPEN, the magnetization phase is a parabolic function in position. Spin-echo SPEN techniques offer high tolerance to B_0 inhomogeneity and are having utility for studying tissues in which variations in magnetic susceptibility are large, like the prefrontal orbital cortex of the human brain [32].

3. Basics of FM pulses

A rotating reference frame (x′y′z′) provides the best platform from which to visualize the motion of a magnetization vector **M** experiencing the torque from a magnetic field vector **B**. In a frame rotating about $\mathbf{B_0}$ at the angular velocity ω_{RF} of $\mathbf{B_1}$, the on-resonance condition occurs when the RF frequency is equal to the spin's resonant frequency; i.e., when $\omega_{RF} = \omega_0$ (Fig. 1a). An important feature of any RF pulse is how uniformly it rotates **M** when a frequency offset $\Delta\omega$ occurs, as for example in the presence of a field gradient used for slice selection. In the off-resonance case, the axis of rotation is tilted out of the transverse plane (Fig. 1b).

Usually, the pulse shapes used in MRI and MRS are amplitude-modulated (AM). The most common examples include shapes defined by sinc and gauss functions. These originated in the early days of MRI and were derived from a Fourier transform (linear) approximation to the Bloch equations. The frequency offset ($\Delta\omega$) range over which a pulse rotates the magnetization is known as the pulse bandwidth, b_w. The bandwidth is inversely proportional to the pulse duration T_p and depends on the specific pulse pattern (e.g., sinc versus gauss) and the flip angle that is used. The latter is a consequence of the non-linearity of the Bloch equations.

With AM pulses, the carrier frequency (ω_{RF}) remains constant during RF irradiation. With FM pulses, the pulse can be both amplitude and frequency modulated. This difference is illustrated in Fig. 2. Implementation of FM pulses on MR scanners usually requires modulating the pulse phase, instead of modulating the pulse frequency. Comparisons of the amplitude and phase functions ($B_1(t)$ and $\phi(t)$) and slice profiles of two AM pulses and an FM pulse are shown in Fig. 3. The slice profiles produced by these pulses were obtained from Bloch simulations using the pulse parameters listed in Table 1. Slice profiles as a function of the peak of the B_1(t) function (B_1^{max}) are shown in Fig. 4. It can be seen that the FM pulse produces the same π rotation when the RF amplitude ($\omega_1^{max} = \gamma B_1^{max}/2\pi$) meets or exceeds the threshold value that satisfies the adiabatic condition.

With an FM pulse, ω_{RF} is time dependent, and therefore, the amplitude of $\Delta\omega\hat{\mathbf{k}}$ and the amplitude and orientation of the effective field $\vec{\boldsymbol{\omega}}_{eff}$ change during the pulse. Here we briefly describe the motions of the time-dependent magnetization and the field components in a rotating frame, in response to an FM pulse. At any moment during the pulse, the rate at which $\vec{\boldsymbol{\omega}}_{eff}(t)$ changes its orientation is given by the instantaneous angular velocity, $d\alpha(t)/dt$, where α is the angle between $\vec{\boldsymbol{\omega}}_{eff}$ and the z′-axis. At the beginning of the pulse ($t = 0$), if $\Delta\omega \gg 0$, then the magnitude of $\Delta\omega\hat{\mathbf{k}}$ is very large relative to that of $\omega_1\hat{\mathbf{i}}$, and thus, the initial orientation of $\vec{\boldsymbol{\omega}}_{eff}$ will be approximately collinear with z′. As $\omega_{RF}(t)$ increases during the pulse, $\Delta\omega$ decreases and $\vec{\boldsymbol{\omega}}_{eff}(t)$ rotates toward the transverse plane. When $\omega_{RF}(t) = \omega_0$, the orientation of $\vec{\boldsymbol{\omega}}_{eff}$ is parallel to $\omega_1\hat{\mathbf{i}}$, regardless of the magnitude of the $\omega_1\hat{\mathbf{i}}$. In a classical adiabatic half-passage (AHP), the orientation of $\vec{\boldsymbol{\omega}}_{eff}$ is swept in this manner from z′ to an axis in the transverse plane. In an adiabatic full-passage (AFP), the sweep of $\omega_{RF}(t)$ continues past resonance so that the final orientation of $\vec{\boldsymbol{\omega}}_{eff}$ is parallel with −z′ (i.e., at the end of the AFP, $\Delta\omega \ll 0$). During an adiabatic pulse, a magnetization vector (**M**) which is parallel to $\vec{\boldsymbol{\omega}}_{eff}$ will tend to follow $\vec{\boldsymbol{\omega}}_{eff}$, provided that $|\vec{\boldsymbol{\omega}}_{eff}(t)| \gg |d\alpha(t)/dt|$, for all t. This inequality is known as the "adiabatic condition". In simple terms, the adiabatic condition states that, at all times during the pulse, the rate at which $\vec{\boldsymbol{\omega}}_{eff}$ changes its orientation must be small relative to the rate at which a magnetization vector rotates about $\vec{\boldsymbol{\omega}}_{eff}$. Adiabatic