

**Reduction of Artifacts Arising From Non-Ideal Gradients
in Fast Magnetic Resonance Imaging**

BY

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A DISSERTATION

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To,

my fiancé, Vijay;

our parents,

Maya and Abbasi,

Jaya and Viswanathan;

and our sisters,

Neelam and Divya;

who helped me believe

in myself.

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LIST OF ABBREVIATIONS

ACR	American College of Radiology
APA	Adaptive Phased Array
ASL	Arterial Spin Labeling
B_0	Main magnetic field
B_1	Radiofrequency field
BOLD	Blood Oxygenation Level Dependent
BRM	Body Resonator Module
CPMG	Carr-Purcell-Meiboom-Gill
CRM	Cardiac Resonator Module
CT	Computed Tomography
CTL	Cervical-Thoracic-Lumbar
DICOM	Digital Imaging and COMMunications in Medicine
DSC	Dynamic Susceptibility Contrast
DSI	Diffusion Spectroscopic Imaging
DSV	Diameter of Spherical Volume
DWI	Diffusion Weighted Imaging
DTI	Diffusion Tensor Imaging
emf	electromotive force

LIST OF ABBREVIATIONS (Continued)

EPI	Echo Planar Imaging
ESP	Echo Spacing (also called Spacing between Echoes)
ETL	Echo Train Length
EVI	Echo Volumar Imaging
FID	Fred Induction Decay
FLAIR	Fluid Attenuated Inversion Recovery
FLASH	Fast Low-Angle SHot
FOV	Field of View
FFT	Fast Fourier Transform
FSE	Fast Spin Echo
FT	Fourier Transform
GE	Gradient Echo
GRASE	Gradient and Spin Echo
GRE	Gradient Recalled Echo
HARDI	High Angular Resolution Diffusion Imaging
iFFT	inverse Fast Fourier Transform
LAP	Long-Axis PROPELLER
MR	Magnetic Resonance
MRI	Magnetic Resonance Imaging
MRS	Magnetic Resonance Spectroscopy
NEX	Number of Excitations

LIST OF ABBREVIATIONS (Continued)

NMR	Nuclear Magnetic Resonance
ONG	Oblique Nyquist Ghost
PC	Phase Contrast
PE	Phase-Encoding
PNS	Peripheral Nerve Stimulation
ppm	parts per million
PROPELLER	Periodically Rotated Overlapping Parallel Lines with Enhanced Reconstruction
RARE	Rapid Acquisition with Relaxation Enhancement
RF	Radio-frequency
RO	ReadOut
ROI	Region of Interest
RS	Readout Segmented
SAP	Short-Axis PROPELLER
SAR	Specific Absorption Rate
SCIC	Surface Coil Intensity Correction
SE	Spin Echo
SENSE	SENSitivity Encoding
SNR	Signal-to-Noise Ratio
SPSP	SPatial-SPectral
SS	Slice Selection
SSFP	Steady-State Free Precession

LIST OF ABBREVIATIONS (Continued)

T_1	Longitudinal relaxation time
T_2	Transverse relaxation time
TMS	TetraMethyl Silane
TR	Repetition Time
TE	Echo Time

SUMMARY

The goal of this PhD research project was to develop and validate two techniques for the reduction of image artifacts that arise from the non-ideal magnetic field gradient in fast magnetic resonance imaging (MRI). The non-ideal gradient characteristics include: (a) gradient non-linearity away from the magnet isocenter, (b) eddy currents as a result of the gradient pulses, and (c) gradient anisotropy as a result of inconsistent eddy currents among the gradient axes. Image artifacts arising from these non-ideal gradient characteristics can severely compromise image quality and adversely affect clinical diagnoses.

The first artifact is a “feather-like” artifact, termed as cusp artifact, which is typically seen as a bright line or a ‘feather’ along the phase-encoding direction on sagittal and coronal fast spin echo (FSE) images of the spine or the knee. This artifact is a wrap-around artifact due to gradient non-linearity coupled with a spatially varying main magnetic field near the end of the magnet bore. A novel technique is proposed to reduce this artifact, in which an FSE pulse sequence is modified to slightly tilt the slice selected by the radiofrequency (RF) excitation pulse away from the slice selected by the RF refocusing pulses. At the edge of the field of view (FOV), the incomplete overlap between the slices selected by the excitation and refocusing pulses effectively reduces the signals from the artifact-prone region. In contrast, the slices overlap substantially within the FOV so that the signals are largely retained.

This slice-tilting technique was implemented on two commercial MRI scanners operating at 3.0 T and 1.5 T, respectively, and evaluated on phantoms and human spine and extremities using clinical protocols. Both phantom and human results showed that applying the technique decreased the intensity of the artifact by 50~90% of the original, and substantially limited the spatial extent of the artifact. This

SUMMARY (Continued)

technique can be implemented on most MRI scanners without hardware modification, complicated calibration, sophisticated image reconstruction, or patient-handling alteration.

The second artifact is a Nyquist ghost artifact occurring in a k -space sampling scheme known as “PROPELLER” that is based on an echo-planar imaging (EPI) pulse sequence. The Nyquist ghost is primarily due to the constant and linear phase errors caused by eddy currents due to the readout gradient. Specific to EPI-PROPELLER pulse sequences, gradient anisotropy can cause another type of phase errors, which are called oblique phase errors. This type of phase errors leads to a different Nyquist ghost, known as oblique Nyquist ghost (ONG). A time-efficient phase correction technique has been developed to simultaneously correct constant, linear, and oblique phase errors in short-axis and long-axis EPI-PROPELLER sequences. This technique relies on the acquisition of two or three reference scans along orthogonal physical gradient axes. Constant and linear phase errors (along the readout direction) were obtained from these reference scans, and then used to predict phase errors for all blades, regardless of blade orientation. Additionally, oblique phase errors were also calculated from the reference scans and subsequently corrected along the phase-encoding direction for oblique blades.

This phase-correction technique was evaluated on both short- and long-axis EPI-PROPELLER images acquired from phantoms and human subjects. With constant and linear phase corrections alone, the phase correction technique reduced the Nyquist ghost to 1.0~2.5% for each blade, when gradient anisotropy effects were negligible. In the presence of significant gradient anisotropy, the phase correction technique reduced the ONG to 0.5~5.0 % for obliquely oriented blades. The ability to simultaneously and time-efficiently reduce Nyquist ghosts arising from three different phase errors is expected to improve the robustness of EPI-PROPELLER sequences particularly on a scanner with residual eddy currents and gradient anisotropy.

1 Introduction

1.1 Background

A recent survey of prominent internists at leading medical institutions across the USA concluded that magnetic resonance imaging (MRI) and computed tomography (CT) ranked first among a list of the most significant innovations in healthcare within the last quarter of the twentieth century (1). This is not surprising; MR imaging has significantly broadened the landscape of diagnostic imaging capabilities since the concept was first demonstrated in 1973 (2,3). Today, MRI is widely used due to its excellent soft tissue contrast, its high resolution capability, and the means to acquire images non-invasively and without the use of ionizing radiation.

The basis of MR imaging lies in the phenomenon of nuclear magnetic resonance (NMR). Research in this field was initiated by Isidor Rabi (4) through his experiments to measure nuclear magnetic moments, which later won him the 1944 Nobel Prize in Physics. The NMR signal was measured by Bloch (5) and Purcell (6) in 1945, for which they were jointly awarded the Nobel Prize in Physics, in 1952. Soon after, Bloembergen, Purcell and Pound discovered NMR relaxation effects (7). Although the chemical shift effect was observed in these experiments, it was initially treated as an artifact or technical flaw. It was only after the chemical shift effect was systematically detected (8,9) and understood (10) that NMR spectroscopy was possible. The first resolved spectrum was documented in 1951 (11).

For a long time, nuclear magnetic resonance was regarded as a tool for the chemist alone. The first of two seminal changes to this perception occurred when it was shown that NMR relaxation times differed in healthy and diseased tissue (12). This observation excited the interest of the scientific community in using NMR to study disease. Then, in 1973, Lauterbur (2) and Mansfield (3) independently suggested the use of applied magnetic field gradients to create two-dimensional MR images. It is this

achievement that eventually led to the growth of MRI as a diagnostic imaging tool. Although, MRI was initially viewed as an adjunct technique to CT, which had also recently been developed (13) as a diagnostic tool, it quickly became an imaging modality in its own right, earning Drs. Lauterbur and Mansfield the Nobel Prize in Medicine, in 2003. The pace of development in NMR was also accelerated by Richard Ernst's work on pulsed-wave Fourier transform NMR (14), for which he was awarded the 1991 Nobel Prize in Chemistry. The first commercial MRI scanner was manufactured in 1980. Incidentally, it was the use of the NMR phenomenon toward diagnostic imaging of the human body that eventually prompted the community to drop the word 'nuclear', which had begun to acquire negative connotations in the vernacular.

In the thirty years since it was first commercialized, MRI has gone through technological evolution after evolution. Image signal-to-noise ratios (SNRs) have increased due to improvements in hardware, the most important of them being the progressive increase in the strength of the main magnetic field (B_0). Gradient field strengths have also increased, allowing higher-resolution images to be acquired within clinically acceptable scan times. Advances in magnetic field gradient technology (higher field strengths and faster slew rates) have also led to the development of fast acquisition techniques (15), which have improved patient throughput considerably. For example, a T_2 -weighted set of 42 images (with whole brain coverage) takes eighteen minutes to acquire using the spin echo sequence, but can be acquired within five minutes using the fast spin echo (FSE) sequence, with very similar quality. The use of parallel imaging (16) has further increased the speed of data acquisition while largely maintaining image quality. More recently, parallel transmission (17,18) of the radiofrequency (RF) field (called ' B_1 field') has been made possible, allowing the creation of more uniform radiofrequency excitation fields with less RF power deposition.

These technological advances have given rise to new research and clinical applications of MRI. MRI is now routinely used before neurosurgery (19,20) to identify neurologically damaged areas from tissue that can be salvaged. Functional neuroimaging (21) has also added several dimensions to our

understanding of human cognitive processes. Diffusion and perfusion imaging are powerful tools to evaluate ischemia (22). Real-time anatomical imaging of moving structures, such as the heart (23), has made it possible to evaluate physiological changes and abnormalities non-invasively. Many of these advances have been possible due to fast pulse sequences, resulting in clinically viable scan durations. All in all, technological advances in MRI have helped to shape our understanding of physiology and pathology.

1.2 Problem Definition and Specific Aims

Advances in MRI hardware and technology usually go hand-in-hand with related drawbacks. For example, as field strengths have increased, magnetic field homogeneity over the region of interest has become harder to maintain. The result is non-uniform signal intensity across the field of view (FOV) of interest, which can be corrected using a number of approaches (24,25). Similarly, higher-amplitude gradient fields that allow higher spatial resolution and faster imaging capability also come with their own limitations. These limitations are caused by the non-ideal nature of the gradient field, and are present to some degree in all MRI scanners. Among these limitations are (a) gradient non-linearity away from the center of the magnet bore (called ‘isocenter’), (b) the formation of eddy currents which change the amplitude of the net gradient field, and (c) inconsistent eddy current characteristics and delays between the gradients along the x , y , and z axes.

Image artifacts can arise from each of the non-ideal gradient behaviors, particularly when fast imaging pulse sequences are employed. This is not only because fast sequences rely on the gradient technology, but also because many of these sequences consist of echo trains, resulting in error accumulation over the data acquisition time. Examples of such artifacts include the cusp artifact (26) seen in the fast spin echo (FSE, (27)) pulse sequence, and the ubiquitous Nyquist ghost (28) seen on all echo planar imaging (EPI, (29))-based pulse sequences. These artifacts can severely reduce image quality and thus compromise clinical diagnoses. As commercial MRI scanner manufacturers introduce gradient

systems capable of higher fields and slew rates, these non-ideal gradient behaviors can increase the severity of image artifacts.

It is important to develop robust techniques for the reduction of imaging artifacts caused by non-ideal gradient behavior. This PhD project explored the reduction of two artifacts: the FSE cusp artifact, and the Nyquist ghost in EPI-based PROPELLER (periodically rotated overlapping parallel lines with enhanced reconstruction, (30)) sampling strategies. The first artifact – the FSE cusp artifact – is typically seen as a bright line or a ‘feather’ along the phase-encoding direction on sagittal and coronal FSE images of the spine and knee (31). This artifact is a wrap-around artifact arising from gradient non-linearity coupled with a spatially varying main magnetic field near the end of the magnet bore. The second artifact is a Nyquist ghost artifact occurring in a k -space sampling scheme known as “PROPELLER” that is based on an echo-planar imaging (EPI) pulse sequence. The Nyquist ghost is a manifestation of the constant and linear phase errors caused by eddy currents due to the readout gradient. Specific to oblique imaging planes (such as in EPI-PROPELLER pulse sequences), gradient anisotropy (32) can cause another type of phase errors (33), which we call “oblique phase errors”. This type of phase errors leads to a different Nyquist ghost, known as oblique Nyquist ghost (ONG). The aim of this PhD project was to explore the formation of these two artifacts and to propose a unique, intuitive, and robust approach for their reduction. To that end, the following specific aims were proposed:

1. *Aim 1:* To develop and validate a technique to reduce the FSE cusp artifact by at least 50%. This is accomplished by tilting the slice selected during the RF excitation pulse by a small angle. This technique has been implemented on MRI scanners operating at 1.5 T and 3.0 T, and was validated on phantoms and human subjects. Further, the slice-tilting approach was proposed as a method to reduce aliasing artifacts in the acquisition of non-axial diffusion-weighted PROPELLER images. This technique was implemented and validated on a human subject at 1.5 T.
2. *Aim 2:* To develop and validate a unified phase-correction technique to reduce the Nyquist ghosts arising from eddy currents and gradient anisotropy in EPI-PROPELLER images to 0.5~5%. The

technique has been validated on short- and long-axis EPI-PROPELLER images acquired in several imaging planes, on a 3.0 T MRI scanner on phantoms and human subjects. This technique has further been validated on diffusion-weighted EPI-PROPELLER images.

1.3 Significance of the Project

Fast MRI sampling techniques (15), such as EPI, FSE, and PROPELLER, have been developed as a means to acquire images rapidly while maintaining image quality. However, strong and rapidly switching gradient fields used in fast imaging techniques result in artifacts that can severely degrade image quality, adversely affecting clinical diagnoses. The development of the proposed artifact reduction techniques will make FSE and EPI-PROPELLER more robust in clinical and research applications. Additionally, the proposed techniques can serve as prototypes for further development or implementation of artifact-reduction schemes in fast MRI.

The slice-tilting technique proposed in *Aim 1* has immense potential for clinical imaging as the reduction of the strength and width of the cusp artifact will not only improve image quality but also make it possible to distinguish anatomical features that otherwise may have been obscured by the artifact. Also, engineering trends in magnet design are moving towards shorter bores and higher maximum gradient field strengths. The combination of these two factors will move the artifact-prone region closer to the magnet isocenter, making it necessary to apply robust, clinically-viable techniques for the correction of such peripheral signal artifacts.

In addition to its primary role in the reduction of the FSE cusp artifact, slice tilting has been proposed to reduce aliasing artifacts in reduced FOV imaging (34). With this application in mind, the proposed technique was also applied to non-axial FSE-based PROPELLER acquisitions, where, due to the changing phase-encoding direction, aliasing can occur if the object extends beyond the prescribed

FOV in any direction. The proposed technique is one among a few recently proposed approaches (35,36) to extend the application of PROPELLER to whole-body anatomical imaging.

Although EPI-PROPELLER offers a significant advantage in acquisition efficiency over FSE-based PROPELLER, it also faces several challenges to its widespread application, partly due to gradient hardware-related phase-errors. In EPI-PROPELLER, the phase errors differ with each blade, necessitating blade-specific correction, which can be time-intensive or computationally intensive. The phase correction problem in EPI-PROPELLER is complicated by the fact that non-orthogonal blades can exhibit the oblique Nyquist ghost, which cannot be corrected with the standard phase correction technique used by most clinical scanners. The phase-correction technique proposed in *Aim 2* not only unifies constant, linear *and oblique* phase correction within a single scheme, but also facilitates a more efficient utilization of MRI scanning time. The phase-correction algorithm can also be applied to oblique scanning planes, potentially advancing the application of all EPI-based sequences to anatomically challenging regions, such as the heart and spine, where arbitrary imaging planes are routinely selected.

1.4 Thesis Organization

In this thesis, two imaging artifacts seen in fast pulse sequences will be investigated, and a scheme for the reduction of each artifact will be proposed and validated. These two artifacts are the cusp artifact seen in fast spin echo images, and the Nyquist ghost seen in EPI-based PROPELLER. The chapter-wise organization of this thesis is as follows:

Chapter 2 introduces the principles of nuclear magnetic resonance, including radiofrequency excitation, signal reception, and the principles of the Fourier transform. These topics form the basis of NMR spectroscopy, which will also be introduced in this chapter.

Chapter 3 describes the basis of magnetic resonance imaging, beginning with the concept of a linear imaging gradient and the description of an MRI system. This will be followed by a discussion on

gradient echo, spin echo, and a typical pulse sequence based on each of these standard signal acquisition mechanisms. Imaging gradients have also been used to sensitize the spins to flow, motion, diffusion and perfusion. The theory of diffusion imaging will be introduced in this chapter, as a precursor to its application in Chapters 5 and 6. Finally, gradient non-ideal behaviors, such as non-linearity, anisotropy and the formation of eddy currents, will be introduced, followed by the artifacts introduced in MR images as a result.

Chapter 4 describes pulse sequences for fast imaging. In particular, this chapter will focus on fast spin echo (FSE), echo planar imaging (EPI), and the PROPELLER (Periodically Rotated Overlapping Parallel Lines with Enhanced Reconstruction) sampling technique. This chapter will discuss the advantages and drawbacks of each technique, and the possible artifacts. Some variations of each pulse sequence will be discussed as a way to reduce the inherent artifacts or improve the image quality for specific applications.

Chapter 5 focuses on the FSE cusp artifact. First, the origin and formation of this artifact will be explained, followed by a description of the appearance of the artifact as a function of relevant imaging parameters, such as slice thickness and echo train length (ETL). This introduction will be followed by a detailed discussion of the proposed reduction technique, based on tilting the selected slice by a small tilting angle. Finally, the technique will be validated through intensive experimentation on phantoms and human volunteers, and on several MRI scanners.

Chapter 6 focuses on the Nyquist ghost artifact seen in EPI-based PROPELLER. The origin of the ghost will be explored, the issues with its reduction will be discussed, and a time-efficient phase correction technique will be proposed for its reduction. This technique will be validated on anatomical as well as diffusion-weighted EPI-PROPELLER images on axial and oblique imaging planes.

Chapter 7 provides a summary of the project, highlighting original contributions and pointing to future research directions.

2 Nuclear Magnetic Resonance

This chapter introduces the basic mechanisms responsible for the development of the field of nuclear magnetic resonance imaging. The fields of NMR and, by extension, MRI, are based on the interaction of a nuclear spin with an external magnetic field. The word “magnetic” is used to describe the various magnetic fields that are involved in the formation of the image. The word “resonance” refers to the need to match the oscillating radiofrequency field (B_1) to the precessional frequency (ω_0) of a nuclear spin (37). Although many nuclei possess spin, and can therefore be studied using NMR, clinical MRI applications focus on the hydrogen (^1H) nucleus (containing a single proton), as its large natural abundance and sensitivity make it the easiest to detect and observe within the human body.

2.1 Nuclei in the Presence of an External Magnetic Field (B_0)

Nuclear magnetic resonance is a phenomenon that depends on the paramagnetic behavior of atomic nuclei and their response to an electromagnetic stimulus (38). This behavior is exhibited due to *nuclear spin*, the property that makes the nucleus turn around its own axis, which is possessed by nuclei with an odd atomic weight and/or odd atomic number. The amplitude of the spin angular momentum, $|P|$, is given by:

$$|P| = \frac{h}{2\pi} \sqrt{I(I+1)} \quad [2.1]$$

where h is Planck’s constant (6.63×10^{-34} J-s) and I is the spin quantum number of the nucleus, which can only take integral or half-integral values. Since the nuclei also carry an electric charge, the combination of charge and spin makes the nuclei act like tiny magnetic dipoles, with a magnetic moment given by:

$$|\mu| = \gamma |P| \quad [2.2]$$

where γ is a constant called the gyromagnetic ratio, expressed in rad/s/T (1 Tesla = 10^4 Gauss). In water, the hydrogen nucleus (or proton) has a gyromagnetic ratio of ~ 2.68 rad/s/T, which is equivalent to a spin precessional frequency of 42.58 MHz/T.

The spin angular momentum and the magnetic moment vector are oriented at an angle of θ away from the axis of precession of the spin, which is determined by:

$$\theta = \cos^{-1} \left(\frac{m}{\sqrt{I(I+1)}} \right) \quad [2.3]$$

where m has $(2I + 1)$ values, given by $m = I, I - 1, I - 2, \dots, -I$, and is the number of possible energy states that a spin- I system can have. This means that, for the hydrogen nucleus ^1H with $I = \pm 1/2$, and $m = 2$, the magnetic moment vector is oriented $\pm 54.74^\circ$ away from its axis of precession, hereafter referred to as the z-axis. The angular momentum of the spin in the z-direction is therefore given by:

$$P_z = m \frac{h}{2\pi} \quad [2.4]$$

and the component of the magnetic moment along the longitudinal z-axis is:

$$\mu_z = \gamma m \frac{h}{2\pi} \quad [2.5]$$

In the absence of any external electromagnetic field, each nucleus in a spin system aligns itself in a random direction, due to the presence of thermal energy. The net magnetization of an ensemble of spins is therefore zero. In the presence of a strong, external, static magnetic field, B_0 , the nucleus acquires a magnetic energy, given by:

$$E = -\mu_z B_0 = -\gamma m \frac{h}{2\pi} B_0 \quad [2.6]$$

Since m has two possible values, the energy levels, E , are also quantized. For a particle with $I = \pm\frac{1}{2}$, the difference in energy levels is given by:

$$\Delta E = \gamma \frac{h}{2\pi} B_0 \quad [2.7]$$

The angular frequency of an oscillating magnetic field that corresponds to an electromagnetic wave with energy ΔE is given by:

$$\omega_0 = \frac{\Delta E}{h} \quad [2.8]$$

From Eqs. 2.7 and 2.8, we can derive the relationship:

$$\omega_0 = \gamma B_0 \quad [2.9]$$

The frequency, ω_0 , is called the *Larmor frequency*, and forms the basis of the resonance phenomenon in NMR. In the field of MRI, Eq. 2.9 is commonly referred to as the *Larmor equation*. This derivation relies on the principles of quantum mechanics. It is also possible to derive the Larmor equation using classical physical principles. This has been shown in texts by de Graaf (39) and Haacke, *et al.* (37).

Not all nuclei orient in the direction of the externally applied magnetic field, B_0 . Due to the thermal energy associated with environmental factors such as the absolute temperature (T) of the system, nuclei at higher energy levels can orient away from the direction of the field. For example, the proton (hydrogen nucleus) with a spin of $\pm\frac{1}{2}$ orients in two possible ways: parallel and anti-parallel to the external magnetic field, B_0 . Each possible orientation of the nucleus has an energy level associated with it: the lower energy level associated with the nucleus oriented along the direction of the field, and the higher energy oriented opposite to the field. In the presence of an external field, a net *longitudinal equilibrium magnetization* (M_0) exists in the direction of the field as there are more protons aligned with the field than against it. This resulting magnetization is given by:

$$M_0 = \frac{\rho_0 \gamma^2 \hbar^2}{8\pi^2 kT} B_0 \quad [2.10]$$

where ρ_0 are the protons per unit volume (also called the ‘spin density’) and k is the Boltzmann’s constant (1.38×10^{-23} Joule/K). The excess spins resulting in the net magnetization, M_0 , are a very small number. For example, the excess number of spins aligned parallel to a static magnetic field $B_0 = 3.0$ T is approximately ten out of every 10^6 , at a temperature of 27 °C (300 K). In practice, however, M_0 is observable as the total number of spins is much larger than 10^6 . In general, we are interested in the interactions of large spin systems. The bulk magnetization vector, M_0 , is therefore commonly used in calculations.

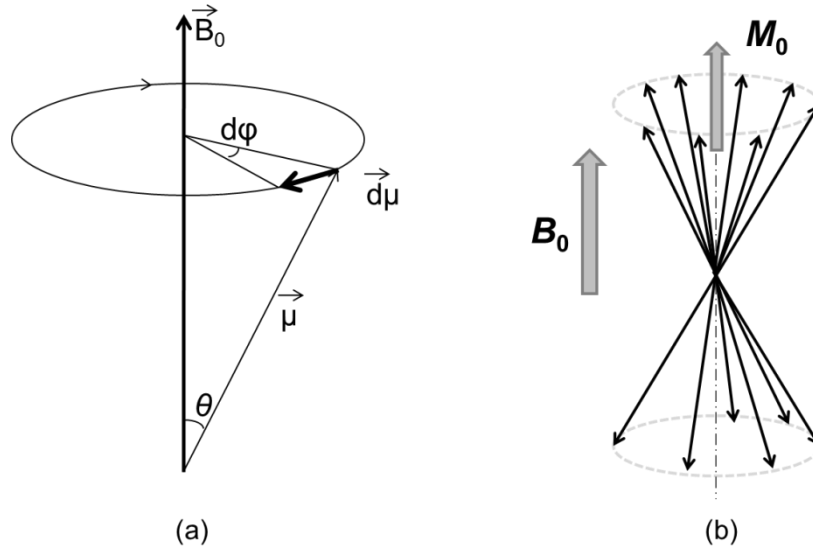


Figure 2.1: (a) The precession of a nucleus in an external magnetic field, B_0 , and (b) the spin excess due to B_0 (adapted from (39) and (40)).

2.2 The Resonance Condition and the Radiofrequency (B_1) Field

Nuclear spins can transition from the lower to the higher energy state through the application of energy given by Eq. 2.7. This energy is applied in the form of a small external magnetic field having amplitude B_1 . When the B_1 field is applied at the Larmor frequency, ω_0 , it is said to be applied *on resonance*. The condition of being on resonance means that the B_1 field is maximally synchronized to tip the nuclear spins from the lower to the higher energy state.

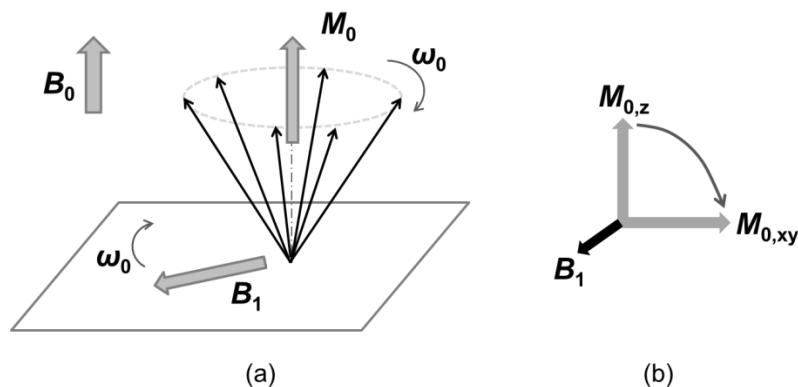


Figure 2.2: A circularly polarized RF pulse, illustrated in (a), in the laboratory frame of reference, and its effect on the net magnetization, shown in (b), in the rotating frame of reference (adapted from (40)).

A B_1 field applied on resonance is called an RF pulse. Figure 2.2a shows the transverse B_1 field rotating synchronously with the precessing spins. When this *circularly polarized* RF field, given by:

$$B_{1cir}(t) = B_1(\cos \omega_0 t \cdot \hat{x} + \sin \omega_0 t \cdot \hat{y}) \quad [2.11]$$

is directed perpendicularly to the main magnetic field, the net magnetization, M_0 , nutates through the desired angle, α , given by:

$$\alpha = \gamma \int_0^{\tau} B_{1cir}(t) \cdot dt \quad [2.12]$$

where τ is the duration of the RF pulse. For the purpose of the following discussion, we denote $\hat{x}$, $\hat{y}$ and $\hat{z}$ as unit vectors in the Cartesian co-ordinate system. If the B_1 field rotates the net magnetization by 90° , all of the initial longitudinal magnetization ($M_{0,z}$) is converted into transverse magnetization ($M_{0,xy}$), as shown in Figure 2.2b. The bulk action of the spins can be understood by using a *rotating frame of reference*, where the coordinate system is rotating at the Larmor frequency. Such a frame of reference simplifies the description of the motion of the magnetization vector and the RF field over the stationary or *laboratory reference frame*. The rotating reference frame will be used as the basis for all future discussions in this thesis.

When the RF pulse is removed, the spins eventually come back to rest along the direction of the main magnetic field. The behavior of the spins at this time depends upon their interaction with each other, and with the environment of atoms surrounding them (called ‘lattice’). This interaction occurs along with the exchange of energy, and modulates the net magnetization. The mechanisms of spin interaction are discussed in the following section.

2.3 The Bloch Equation and Relaxation in Tissue

The magnetic moment of spins in an external magnetic field, B_0 , tend to align themselves such that they are along the magnetic field. This occurs due to the tendency of the spins to have the least potential energy associated with the external field. Since the spins are in thermal contact with the lattice

formed by nearby atoms, the thermal motion present in the lattice can account for any change in the energy of a given spin (or protons, as in this case). In other words, the spin can exchange energy with the lattice, and, in doing so, return more quickly to its equilibrium position along the external field. This action rebuilds the longitudinal magnetization at time t , $M_z(t)$, to its equilibrium value (M_0), at the same time depleting the transverse magnetization, $M_{xy}(t)$. The rate of change of the longitudinal magnetization is proportional to the difference ($M_0 - M_z$), and can be expressed using the ‘spin-lattice relaxation time’, T_1 , as follows:

$$\frac{dM_z}{dt} = \frac{1}{T_1}(M_0 - M_z(t)) \quad [2.13]$$

T_1 is defined as the time taken for the longitudinal magnetization to reach 63% of the maximum possible signal, M_0 . After the RF pulse is applied, the repletion of the longitudinal magnetization follows the equation:

$$M_z(t) = M_z(0)e^{-t/T_1} + M_0(1 - e^{-t/T_1}) \quad [2.14]$$

which reduces to only the second term when the flip angle $\alpha = 90^\circ$.

Table I: T_1 and T_2 relaxation times for different tissues at 3.0 T (Adapted from (41))

Tissue	T_1 relaxation time (ms)	T_2 relaxation time (ms)
Gray matter	1331	85
White matter	832	70
Cerebrospinal fluid	4000	2200
Fat	380	133
Arterial blood	1664	165

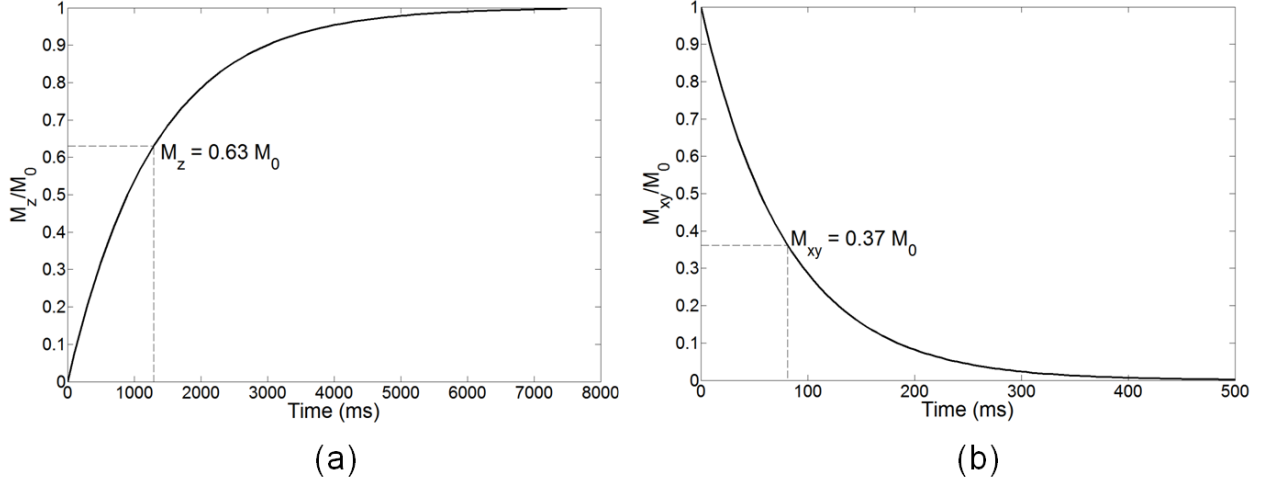


Figure 2.3: (a) Longitudinal (assuming $T_1 = 1300$ ms), and (b) transverse (assuming $T_2 = 80$ ms) relaxation of spins in gray matter at 3.0 T.

The relaxation time, T_1 , ranges from tens to thousands of milliseconds for protons in tissue, and increases with increasing field strength. Figure 2.3a shows the signal repletion due to longitudinal relaxation in gray matter at 3.0 T, assuming $T_1 = 1300$ ms and $\alpha = 90^\circ$. Table I shows the longitudinal relaxation times for different tissues at 3.0 T.

As mentioned earlier, the transverse magnetization experiences a decrease after the RF pulse is withdrawn. Although this decrease occurs concurrently with the increase of the longitudinal magnetization, the most significant reason for the decrease is the formation of local fields which are caused as a combination of the applied field with the fields produced by the neighboring spins. These local spins lead to local precessional frequencies, which cause the spins to fan out with time. The time-constant with which this *dephasing* occurs is called the ‘spin-spin relaxation time’, and can be used to characterize the depletion of the transverse magnetization at time t , $M_{xy}(t)$, using the following equation:

$$\frac{dM_{xy}}{dt} = -\frac{M_{xy}(t)}{T_2} \quad [2.15]$$

with its solution given by:

$$M_{xy}(t) = M_{xy}(0).e^{-t/T_2} \quad [2.16]$$

The relaxation constant T_2 is defined as the time taken for the signal to decay to 37% of its initial value, M_0 . This time is in the order of tens of milliseconds for protons in most tissues, and is much lower than T_1 for the same tissue, because the spin-spin relaxation is a combination of dephasing and the spin-lattice relaxation discussed earlier. T_2 values are much shorter for solids and much longer for liquids. Figure 2.3b shows the signal decrease due to transverse relaxation in gray matter at 3.0 T, assuming $T_2 = 80$ ms. Table I shows the transverse relaxation times for different tissues at 3.0 T.

There is an additional source of spin dephasing that arises due to the inhomogeneity of the external magnetic field, and can be expressed using the time constant:

$$T_2' \propto \frac{1}{\gamma \Delta B_0} \quad [2.17]$$

The total transverse relaxation time constant is given by:

$$\frac{1}{T_2^*} = \frac{1}{T_2} + \frac{1}{T_2'} \quad [2.18]$$

The loss of signal due to T_2' decay is recoverable, and the mechanism for its recovery is discussed in Section 3.4. Intrinsic T_2 losses are related to local, random, and time-dependent variations, and are not recoverable.

The differential equations given by 2.13 and 2.15, for longitudinal and transverse relaxation, respectively, can be combined with the equation for magnetization in the presence of an external magnetic field, into the combined Bloch equation:

$$\frac{d\vec{M}}{dt} = \gamma \vec{M} \times \vec{B}_{ext} + \frac{1}{T_1} (M_0 - M_z(t)) \hat{z} - \frac{1}{T_2} \vec{M}_{xy}(t) \quad [2.19]$$

where $\vec{B}_{ext}$ is the external magnetic field. If $\vec{B}_{ext} = B_0 \hat{z}$ is the static B_0 field along the z -direction, the solution to the Bloch equations is given by Eqs. 2.14 and 2.16. The Bloch equations can also be solved when the external magnetic field $\vec{B}_{ext} = B_0 \hat{z} + B_1 \hat{x}$ is a combination of the B_0 field in the z -direction and the RF field (B_1) in the x -direction.

2.4 Signal Detection and the Fourier Transform

After the application of the RF excitation pulse, the magnetization is tipped completely or partially into the x - y plane. The magnetization precesses about B_0 with the Larmor frequency and induces an electromagnetic force (emf) in the receiving coil which is placed in a plane transverse to the B_0 field. The emf decreases as a function of time due to a combination of relaxation processes and the microscopic and macroscopic inhomogeneity in the main magnetic field. The time dependence of the emf (or received signal) is called *free induction decay* (FID). The x and y components of the transverse magnetization are detected separately and given by:

$$\begin{aligned} M_x(t) &= M_0 \cos[(\omega_0 - \omega)t + \phi] e^{-t/T_2^*} \\ M_y(t) &= M_0 \sin[(\omega_0 - \omega)t + \phi] e^{-t/T_2^*} \end{aligned} \quad [2.20]$$

respectively, where M_0 is the net magnetization along B_0 , ω_0 is the Larmor frequency, ω is the laboratory frequency of the RF pulse, and ϕ is the phase of the signal. $M_x(t)$ and $M_y(t)$ are usually referred to as the real and imaginary FIDs, respectively. These time-domain FID signals are digitized at a high sampling rate. Although the FIDs hold relevant information, such as relative abundance and resonance frequencies, they are usually converted into frequency-domain spectra through a discrete Fourier transformation, before interpretation and post-processing.

The *Fourier transform* (FT)(42) is a mathematical operation that yields the spectral content of a signal, and is the single most important tool in the reconstruction of magnetic resonance signals. This is because the transverse magnetization can be described very naturally by the FT. The transverse magnetization signal is a sum of the signals from many spins. In Fourier theory, any smooth mathematical function, $f(x)$, can be represented as a combination of sines and cosines, as follows:

$$f(x) = C + \sum_{n=1}^{\infty} a_n \cos(nx) + \sum_{n=1}^{\infty} b_n \sin(nx) \quad [2.21]$$

where C is a constant (or ‘DC’ term), and a_i and b_i are the constants associated with the i th term of the cosine and sine series, respectively. This is called a Fourier series. The approximation is closer as the number of terms increases. The Fourier transform, $F(\omega)$, of a function $f(t)$ can be thought of as similar to the spectrum of that function’s Fourier series, and is given by:

$$F(\omega) = \frac{1}{\sqrt{2\pi}} \int_0^{\infty} f(t) e^{-i\omega t} dt \quad [2.22]$$

The two functions, $F(\omega)$ and $f(t)$, are called Fourier transform pairs. It is possible to reconstruct $f(t)$ from $F(\omega)$ in an operation called the *inverse Fourier transform*, as follows:

$$f(t) = \frac{1}{\sqrt{2\pi}} \int_0^{\infty} F(\omega) e^{i\omega t} d\omega \quad [2.23]$$

The actual signal is sampled in discrete steps, so a discrete Fourier transform is used in practice. This can lead to aliasing artifacts, which are usually avoided by satisfying the Nyquist criterion, which states that the sampling frequency must be at least twice of the bandwidth of the signal. The signal is sometimes filtered after acquisition to further suppress the aliases.

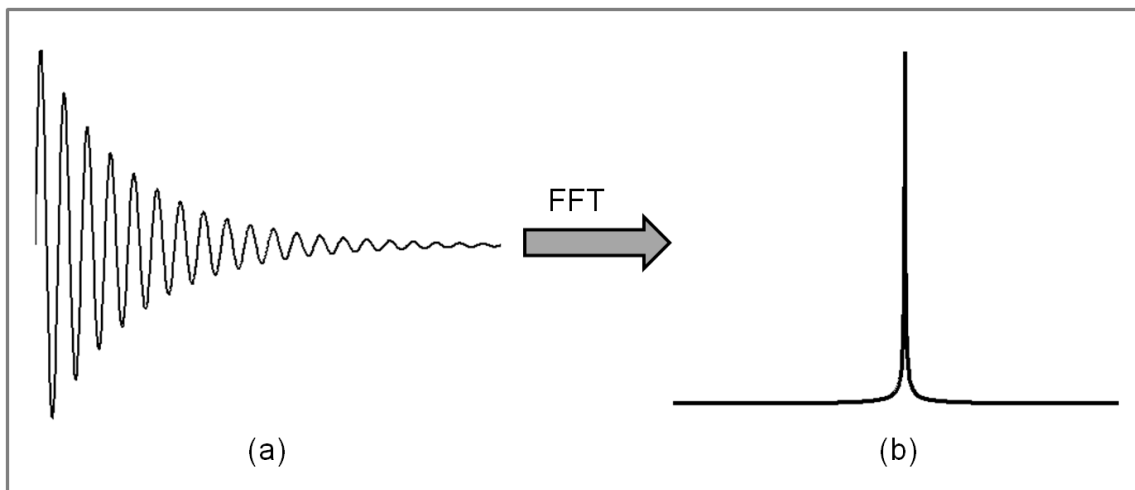


Figure 2.4: (a) the detected signal, and (b) its Fourier transform

As an example, consider a test-tube of pure water. Assuming that there are no local field inhomogeneities, the time-domain FID is shown in Figure 2.4a, and its spectrum after Fourier transform is shown as a single component in Figure 2.4b. In practice, going from the FID to the spectrum involves several other steps, including apodization, phase correction and baseline correction (39). Fourier transform NMR was first demonstrated by Richard Ernst in 1966 (14), for which he received the Nobel Prize in Chemistry in 1991.

2.5 Magnetic Resonance Spectroscopy

The description of nuclear magnetic resonance has, so far, assumed a uniform sample containing only one type of spin, and influenced only by the Larmor frequency in Eq. 2.9. However, the resonance frequency of a nucleus depends not only on the external magnetic field but also on the chemical environment surrounding the nucleus. When a sample containing nuclei with non-zero spin is placed in an

external magnetic field, the electrons surrounding the nucleus slightly shield the nucleus from the magnetic field. The nature and extent of this shielding depends on the nature of the chemical bond and upon the placement of the nuclei within the molecule that it is a part of. The effective field around the nucleus is then given by:

$$B_{eff} = (1 - \sigma)B_0 \quad [2.24]$$

where σ is called the shielding or screening constant, which is a dimensionless number expressed in parts per million and depends upon the chemical environment of the nucleus.

Since there is a direct relationship between the effective magnetic field strength and the Larmor frequency, the precise resonant frequency of a nucleus will depend on its local chemical environment (43). Thus there will be a small shift in resonant frequency, which is called *chemical shift*, and is defined by:

$$\delta = \frac{\omega - \omega_{ref}}{\omega_{ref}} \times 10^6 \quad [2.25]$$

where ω and ω_{ref} are the precession frequencies of nuclei in the compound under investigation and in a reference compound, respectively. The reference compound is ideally one that is chemically inert and produces a strong signal well separated from the other resonances. For example, in proton NMR spectroscopy, the reference compound is tetramethyl silane (TMS). The chemical shift is defined in units of parts per million and enables one to distinguish between different compounds or isotopes using NMR. In MR imaging, the concept of chemical shift is most commonly used to preferentially excite (or suppress) signals from fat or water, using spectral selection.

The mechanism of chemical shift forms the basis of NMR spectroscopy. As an example, Figure 2.5 shows the spectrum obtained from a uniform phantom at 3.0 T. The metabolite distribution in this spectrum, primarily consisting of n-acetyl aspartate (NAA, 2.02 ppm), creatine (Cr, 3.02 ppm), choline (Cho, 3.22 ppm), lactate and lipids (Lac + lip., ~1.3 ppm), and glutamate and glutamine (Glx, 2.6~2.8

ppm), is similar to that present in human brain tissue. The amplitude and distribution of the nuclei corresponding to each metabolite are usually in arbitrary units, plotted with reference to their chemical shift (in ppm).

In general, NMR spectroscopy is not a very sensitive technique as the difference in resonant frequency between nuclei is very small. The acquired spectrum is typically the average of several readings, a technique which increases the signal to noise ratio, and correspondingly increases scan duration. The signal can be increased further by going to higher field strengths, which also increases resolving power between metabolites. NMR signals are further improved by using an operation called shimming (25) to make the main magnetic field as homogeneous as possible. These and other technical aspects of NMR spectroscopy are fully discussed in (39); and the clinical aspects are covered comprehensively in (44).

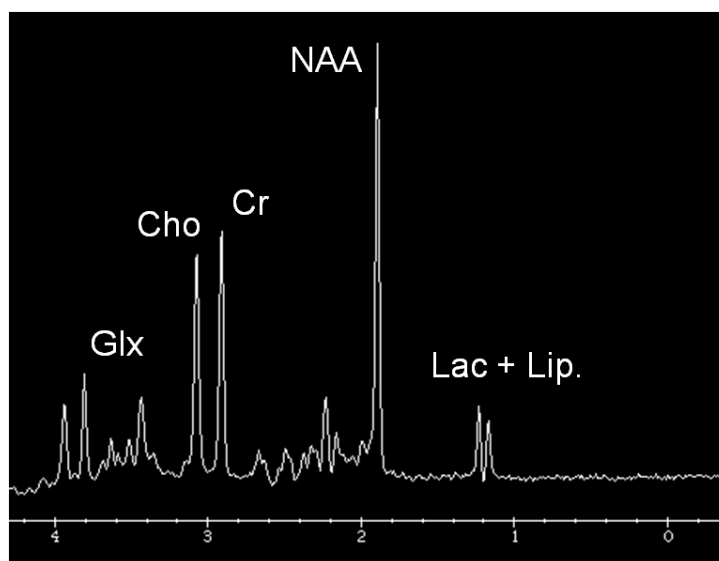


Figure 2.5: MR spectrum from a phantom containing the metabolites present in the brain. The spectral amplitudes (in arbitrary units) are plotted against chemical shifts (in ppm).

3 Magnetic Resonance Imaging

Chapter 2 discussed the principles of nuclear magnetic resonance, specifically focusing on the excitation, reception, and decay of the NMR signal. The two primary hardware elements of such a system are the large static magnetic field (B_0) and small radiofrequency field (B_1), which is turned on and off for signal excitation and reception, respectively. The obtained NMR “spectrum” can give valuable information about chemical structure and composition of the sample. These principles form the basis of NMR spectroscopy.

This chapter discusses the application of the principles of magnetic resonance to the field of imaging (MRI). A magnetic resonance image is formed by using an additional, spatially varying magnetic field, called a ‘*gradient field*’, to differentially encode the spins in the excited region. In most cases, the magnetic field gradient varies linearly in space. The differential encoding resulting from the gradient field makes it possible to obtain a two- or three-dimensional map where the intensity at each point corresponds to the strength of the MR signal detected at that point. This is the MR image. The steps, involving RF excitation, spatial encoding, and data acquisition, are generally played out in order, in a timing diagram referred to as a *pulse sequence*. In this chapter, the process of forming an MR image is discussed in detail, along with the basic pulse sequences required to produce the image.

3.1 The Gradient Field

A linear spatial variation of the main field is produced by introducing a linear magnetic field gradient (2). In MRI, only the spatial component of the field in the z-direction (B_z) is changed, in the following manner:

$$\vec{G} = \frac{\partial B_z}{\partial x} \hat{x} + \frac{\partial B_z}{\partial y} \hat{y} + \frac{\partial B_z}{\partial z} \hat{z} \equiv G_x \hat{x} + G_y \hat{y} + G_z \hat{z} \quad [3.1]$$

where G_x , G_y , G_z are the three orthogonal components of the gradient field $\vec{G}$. The net magnetic field is given by:

$$\vec{B} = (B_0 + G_x x_0 + G_y y_0 + G_z z_0) \hat{z} \quad [3.2]$$

where x_0 , y_0 , and z_0 are the spatial locations away from the isocenter along the x , y , and z directions, respectively. Other non-linear spatially encoding magnetic fields are also gradually being introduced in imaging (45,46).

The three gradients G_x , G_y and G_z , are generated by three independent gradient coils, and are characterized by two important parameters: the maximum attainable amplitude (denoted by h , units in mT/m) and the maximum possible slew rate (denoted by SR , units in T/m/s), which is the largest rate of change of the gradient field per unit time. These parameters depend upon the inductance and resistance of the wire and the maximum allowable voltage generated by the gradient driver associated with that gradient coil.

There are two possible restrictions on the performance of the gradient system. The first is that, although the gradient system may be capable of high strengths and slew rates, these parameters may give rise to peripheral nerve stimulation (PNS) in the human body. PNS is characterized by twitching, tickling, or pain. In order to avoid PNS effects, an additional constraint on the maximal rate of change of the net magnetic field (dB/dt) is also imposed on the system. The second constraint is hardware-related, and imposed on the gradient duty cycle, which is defined as the ratio of the gradient active time to the total duration of the pulse sequence. Gradient coils and amplifiers are subject to thermal heating and can be damaged if the duty cycle exceeds a certain limit. Modern MRI systems usually employ cooling around the gradient coils so that the duty cycle limits can be extended.

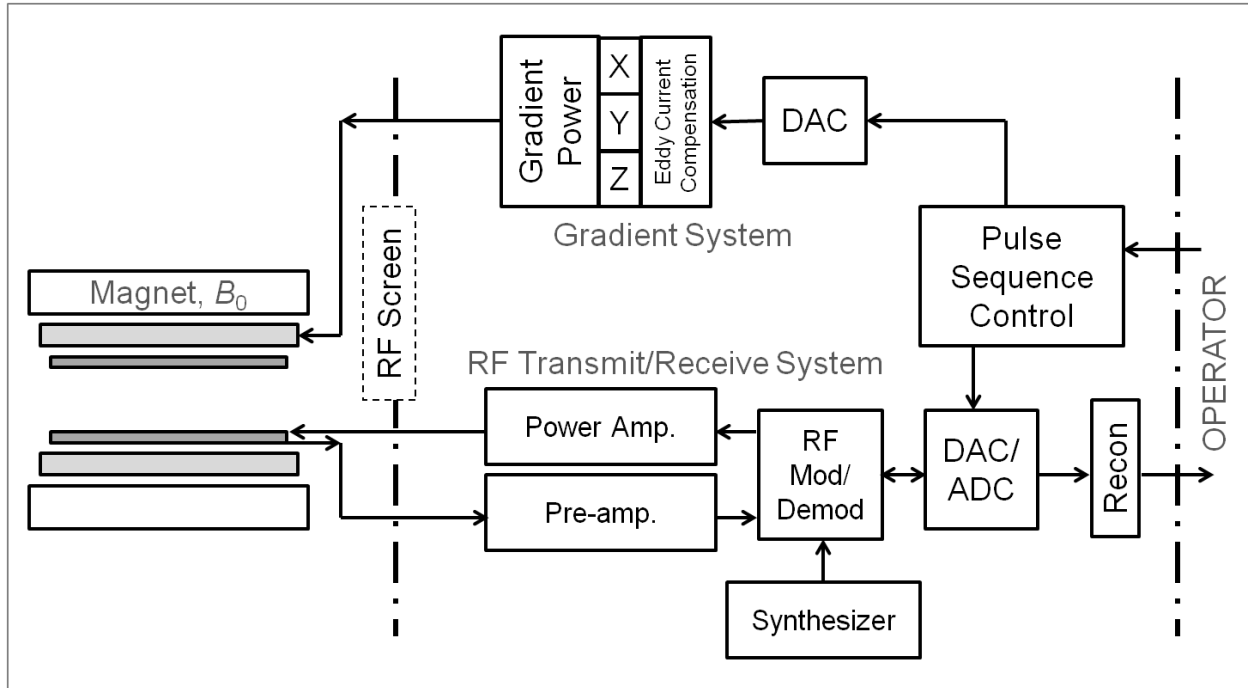


Figure 3.1: Block diagram showing the components of an MRI system.

A block diagram of an MRI scanner is shown in Figure 3.1. This system consists of three primary hardware elements: the main magnet (B_0), the RF transmit/receive system, and the gradient system. After the operator selects the imaging protocol, the parameters are fed synchronously into the RF transmit chain and the gradient system. The first step is the conversion of the digital signal into analog waveforms. The RF modulator receives the signal with frequency of ω_0 from the synthesizer and generates an RF pulse, discussed in Section 2.2, which is amplified before it is sent to the transmit coils to produce the desired B_1 field. Once the RF excitation is applied, the gradient system implements the spatial encoding, which is discussed in further detail in Section 3.2. The transverse magnetization is received by the RF receiver coils (which may be the same or different from the RF transmitter coils), discussed in Section 2.4, and amplified before being demodulated by the carrier frequency (ω_0). As the received signals are very small, an RF screen (made from copper wires) is used to insulate the magnet room from the external

environment, which could introduce spurious variations into the signal. The spatially encoded frequency domain signals are arranged in the form of a two- or three-dimensional map, commonly referred to as *k-space* (see Section 3.3). After the received signals are demodulated and digitized, the image is reconstructed from the *k*-space signals, and displayed to the operator.

3.2 Spatial Encoding

The gradient fields perform three primary functions: slice selection, frequency encoding, and phase encoding. Each of these functions utilizes one or more gradient lobes. A gradient lobe is a single gradient pulse shape that starts and ends with zero amplitude. The pulse sequence consists of a combination of gradient lobes, along each axis, played in a particular order to achieve spatial localization. In general, the following steps are executed: RF excitation with through-plane section selection (called slice selection), frequency encoding and phase encoding for in-plane spatial localization, accompanied by data acquisition. The following sub-sections highlight the role of the imaging gradients for slice selection and spatial encoding.

3.2.1 Slice Selection

A slice selection gradient is always used concurrently with a spatially selective RF pulse, to achieve the desired spatial localization. This gradient is applied perpendicular to the direction dictated by the desired plane. For convenience, this discussion assumes axial slices, where the gradient is applied along the magnet bore, in the physical *z*-direction. The amplitude (G_z) of the slice selection gradient is related to the slice thickness (Δz) and the RF bandwidth ($\Delta\omega_{RF}$) by the following relationship:

$$G_z = \frac{2\pi}{\gamma} \frac{\Delta\omega_{RF}}{\Delta z} \quad [3.3]$$

For a fixed RF bandwidth, the slice thickness is increased by decreasing the amplitude of the slice selection gradient. Figure 3.2 shows the relationship between the slice selection gradient amplitude and the desired slice thickness.

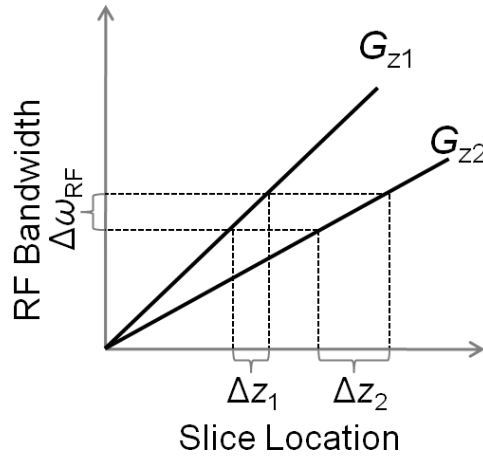


Figure 3.2: The slice selection process, illustrated by a plot of the precession frequency versus the slice location (adapted from (37) and (47)).

The imaging plane can be selected at a location away from the isocenter. In this case, the slice offset from the isocenter, ∂z , is obtained by shifting the carrier frequency of the RF pulse by an amount, $\partial\omega_{RF}$, given by:

$$\frac{\partial z}{\partial \omega_{RF}} = \frac{\Delta z}{\Delta \omega_{RF}} \quad [3.4]$$

This relationship assumes a spatially uniform gradient field. Non-linearity away from the magnet isocenter can cause the selected slice to be distorted, somewhat like a potato-chip. Other spatial

distortions can result from susceptibility variations in the sample, which can perturb the local gradient fields.

The slice-selection gradient, applied concurrent with the RF excitation pulse, can cause phase dispersion of the transverse magnetization across the slice profile. To avoid the signal loss associated with the phase dispersion, a rephasing gradient lobe is usually applied after the slice selection gradient. The slice rephasing lobe is designed to have opposite polarity with respect to the slice selection gradient lobe, and gradient area equal to the area of the slice selection gradient starting from the isodelay point of the RF pulse. Refocusing RF pulses usually do not require rephasing gradients as the phase accumulated in the first half of the refocusing pulse is cancelled during the latter half. In the case of refocusing pulses, however, the slice selection gradient is usually straddled by crusher gradients on either side, which preserve the spin echo while selectively dephasing the FID and stimulated echoes.

3.2.2 *Frequency Encoding*

A linear field gradient can be used to spatially encode NMR signals along the direction of the gradient by assigning a unique precession frequency to each spin isochromat, depending upon its spatial location within the object. The acquired time-domain NMR signals will therefore contain a range of frequencies, which can be revealed through an inverse Fourier transform. The applied gradient is referred to as the frequency encoding gradient, and the process is called *frequency encoding*. As the acquisition of the signal is commonly timed to coincide with the application of this gradient, the frequency encoding gradient is also referred to as the *readout gradient*.

Frequency encoding can be applied along any axis, and typically varies with the type of acquisition. In commonly used pulse sequences, the gradient is applied along a set of parallel lines (commonly called phase encoding, and discussed in Section 3.2.3). This scheme of acquisition is called Cartesian sampling. In non-Cartesian pulse sequences (30,48,49), such as the projection reconstruction

methodology used in CT and originally described for MRI (2), the direction of frequency encoding changes during the course of imaging.

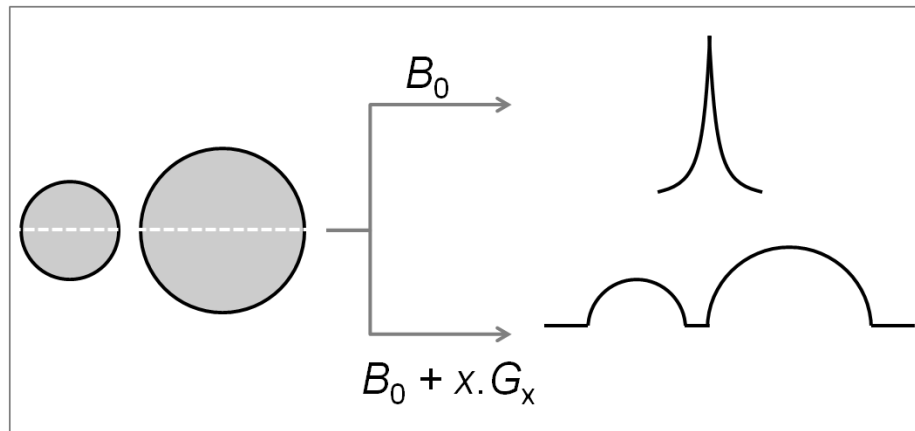


Figure 3.3: Frequency encoding illustrated using a system of two test-tubes filled with water (adapted from (47)).

The effect of the frequency encoding gradient is shown in Figure 3.3. Consider an object consisting of two test-tubes of distilled water separated by a small distance. In the absence of the frequency encoding gradient, the protons in both tubes resonate at the same frequency, producing a single NMR spectrum. When the frequency encoding gradient is applied along the direction shown by the white dotted lines, the resonance frequency of the protons is related to their spatial location along the direction of the gradient. The NMR signal in this case corresponds to a range of frequencies, and the Fourier transform shows the projection of the objects seen through the direction in which the gradient is applied. The amplitude of the readout gradient for each projection is determined by the number of samples in the

projection (N_{FR}), the receiver bandwidth ($\pm\omega_{RO}$), and the selected field of view in the x , or frequency encoding direction (FOV_x), as:

$$G_x = \frac{2\pi}{\gamma} \frac{2\omega_{RO}}{FOV_x} \quad [3.5]$$

where $FOV_x = N_{FR}\Delta x$. Δx is the size of the pixel along the x , or frequency encoding, direction.

The frequency encoding gradient waveforms usually consist of a prephasing gradient lobe and a readout or acquisition gradient lobe. The purpose of the prephasing gradient lobe is to prepare the transverse magnetization in order to create an echo of the signal upon application of the readout gradient. The creation of the echo is discussed in detail in Section 3.4. The echo signal has the largest amplitude when the area of the prephaser is equal to the area of the readout gradient. The echo is acquired symmetrically when the area of the prephasing gradient is half of the area of the readout gradient. It is also possible to design sequences where this relationship is different; in situations where only a fraction of the echo signal is acquired asymmetrically about the echo center (37). If sequences with very short echo time (see Section 3.4) are required, e.g., in lung or bone imaging (50), the prephasing gradient can be entirely dispensed with.

3.2.3 Phase Encoding

In order to form an image, it is necessary to accomplish spatial localization in two dimensions. It is not possible to apply frequency encoding independently in two spatial directions, so phase encoding is used along the second dimension. The phase encoding gradient is usually applied right before the frequency encoding gradient, and along a different direction. In Cartesian imaging, the second direction is chosen orthogonal to the frequency encoding direction. As a result, the linearly varying phase of the MR signal, measured as the angle between the transverse magnetization and an axis (x or y) in the transverse

plane at any time t , is given by: $\phi_y = \gamma y \int_0^T G_y(t) dt$ where $G_y(t)$ is the gradient along the axis (assumed to be y).

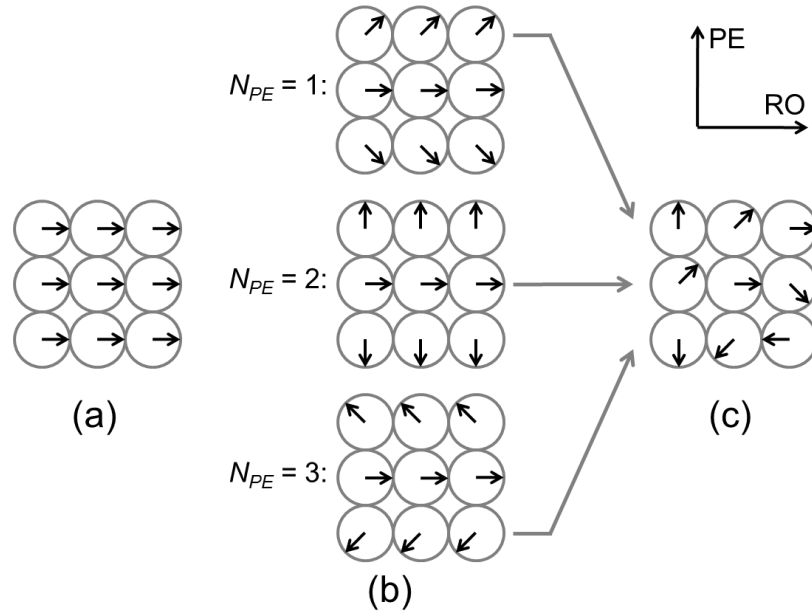


Figure 3.4: Spatial encoding within the selected plane is illustrated using cartoons of the spins (a) after slice selection, (b) upon phase encoding using three phase encoding steps, and (c) after frequency encoding which follows phase encoding. The phase (PE) and frequency (RO) encoding directions are shown on the legend to the right. Relaxation effects are not considered in this figure.

In MRI, the phase encoding gradient is applied several times within an imaging experiment, with varying amplitudes, while keeping the duration of the gradient lobe the same in all cases. As the gradient area changes, the linear phase variation also changes. The linear phase variation is shown schematically in

Figure 3.4. The spins contributing to the transverse magnetization are aligned immediately after the slice selection (Figure 3.4a). Several phase encoding “steps” are needed to acquire the information necessary for the creation of an image (Figure 3.4b). The acquired data contains information that is a composite of both, frequency and phase encoding (Figure 3.4c), and this two-dimensional map is called the k -space map. A two-dimensional MR image can be reconstructed from the acquired information, as discussed in Section 3.3.

For a set of N_{pe} phase encoding steps, the phase encoding step size, Δk_y (i.e., difference between phase encoding steps) must be chosen to satisfy the Nyquist criterion, i.e.,

$$\Delta k_y = \frac{1}{FOV_y} = \frac{1}{N_{PE} \Delta y} \quad [3.6]$$

where FOV_y is the FOV in the phase encoding direction, and Δy is the pixel size along the phase encoding direction. The phase encoding step size determines the difference in areas (and amplitudes) between the gradient lobes for every pair of consecutive phase encoding steps. The maximum phase encoding step size corresponds to the largest gradient area. In order to decrease the echo time, the largest phase encoding step is designed at the maximum gradient strength (h) for that physical gradient axis, as:

$$k_{y,max} = \frac{\gamma}{2\pi} A_{y,max} = \frac{1}{2} (N_{PE} - 1) \Delta k_y \quad [3.7]$$

The maximum gradient area of the phase encoding lobe can be calculated from Eqs. 3.6 and 3.7. The set of phase encoding steps can be executed in sequential order, starting from one end of k -space and moving across the center to the other end. This is usually done in pulse sequences which acquire one line of k -space after every RF excitation. The phase encoding view order is the order in which the phase encoding steps are acquired, and is especially important in fast imaging pulse sequences (see Chapter 4) where several lines of k -space are acquired following every RF excitation pulse. In Cartesian imaging sequences, the phase encoding gradient cannot be played concurrently with the frequency encoding gradient, but it is usually played along with the frequency encoding prephaser, or the slice rephasing gradient. In many

pulse sequences, a phase encoding gradient is followed by a phase rewinding gradient, which is designed with the same area and opposite polarity. The function of the phase rewinding gradient is to rephase the transverse magnetization and make it more consistent between steps.

3.3 Image Reconstruction

Image reconstruction from MRI acquisitions can also be performed using a multidimensional Fourier transform (42). The Fourier transform of a two-dimensional spatial map ($f(x, y)$) is given as follows:

$$F(k_x, k_y) = \frac{1}{2\pi} \int_0^\infty \int_0^\infty f(x, y) e^{-2\pi i(k_x x + k_y y)} dx dy \quad [3.8]$$

where x and y are spatial locations. The two-dimensional k -space map, i.e., $F(k_x, k_y)$ in Figure 3.5a, represents the image in the frequency domain. The k -space parameters, k_x and k_y , are commonly represented in units of cm^{-1} , also called *spatial frequencies*. Similar to the one-dimensional Fourier transform, multi-dimensional Fourier transforms are also invertible. A two-dimensional inverse Fourier transform of the k -space can be used to reconstruct the spatial map, or image $f(x, y)$, in Figure 3.5b as shown:

$$f(x, y) = \frac{1}{2\pi} \int_0^\infty \int_0^\infty F(k_x, k_y) e^{2\pi i(k_x x + k_y y)} dk_x dk_y \quad [3.9]$$

As discussed in Section 2.4, the discrete Fourier transform (51) is used in this case as well. Additionally, most data are acquired or interpolated to powers of two so that the fast Fourier transform (FFT) (51) algorithm can be used for reconstruction.

The Fourier transform is usually the final step in MRI reconstruction. Prior to the Fourier transform, the acquired data may be filtered, phase-corrected, or interpolated to the desired matrix size. This is especially true if only a part of the k -space is acquired (52) or if the acquisition matrix is not a

power of two. Also, if a non-Cartesian acquisition strategy is used, such as spiral (49) or projection reconstruction (2), the data must first be gridded to a standard multi-dimensional matrix before reconstruction using FFT. The discussions on spin and gradient echo pulse sequences assume standard reconstruction practices.

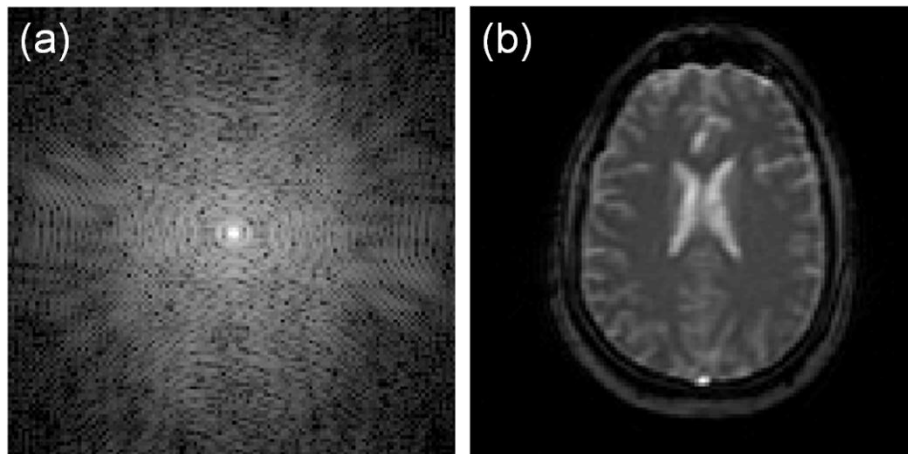


Figure 3.5: (a) k -space, and (b) its corresponding image.

3.4 Spin and Gradient Echoes

Although the free induction decay signal (53) can be used to form an image, it has an extremely short decay time due to magnetic field inhomogeneities, which cause the spin system to dephase very quickly. However, it is possible to use an ‘echo’ signal instead of the FID signal, to acquire images. One of two mechanisms is used to form an ‘echo’ of the signal: spin echo (54) or gradient echo (55,56). Spin echoes are formed when a refocusing RF pulse is applied to the rotating spin system whereas gradient echoes are formed when a combination of prephasing and rephasing gradients (like those used in frequency encoding) are applied.

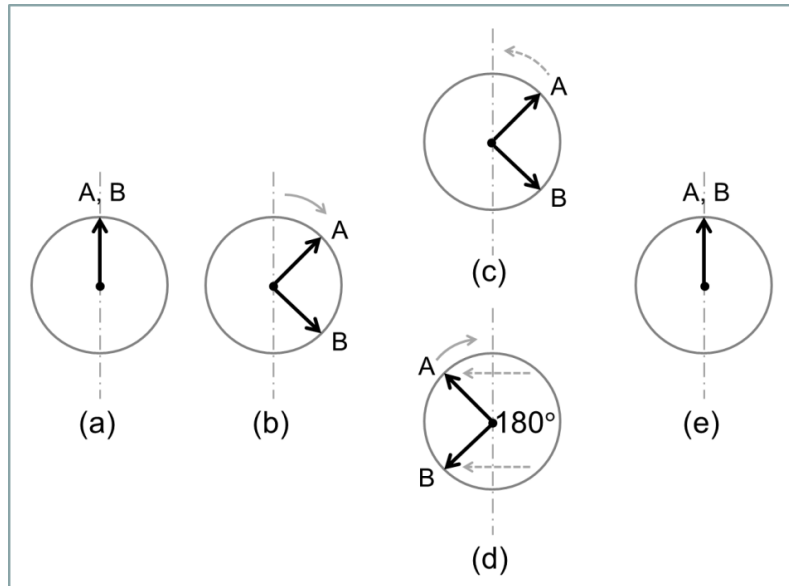


Figure 3.6: illustrates the formation of spin and gradient echoes through a two-spin system shown in the transverse plane. The transverse magnetization is shown (a) immediately after the RF excitation pulse, and (b) at time τ , as a result of the prephasing gradient. In a gradient echo (c), the rephasing gradient with opposite polarity reverses the direction of the spins. In a spin echo (d), the refocusing pulse ‘flips’ the spins. The echo (e) is formed at time 2τ after the excitation pulse. Relaxation effects are not considered in this figure (adapted from (39)).

Figure 3.6 illustrates the formation of spin and gradient echoes. Assume a two-spin system, which is shown in Figure 3.6a, at $t = 0_+$, immediately after the RF excitation pulse is applied. In MRI, a prephasing gradient is applied to spatially encode the spins. After some time, each spin exhibits a position-dependent phase, shown as A and B (Figure 3.6b). At this point, the rephasing (or frequency encoding) gradient is applied (Figure 3.6c). The polarity of the frequency encoding gradient is opposite to that of the prephasing gradient. The effect of this gradient is to reverse the phase accrual (shown by the dotted arrow in Figure 3.6c), such that, when the area of the rephasing gradient is equal to the area of the previously applied prephasing gradient, the spins are back in phase, forming a gradient echo (Figure

3.6e). The time between the application of the RF pulse and the formation of the echo is called the *echo time* (TE). In the gradient echo pulse sequence, the gradients are used to prepare the magnetization, form the echo, and acquire the spatially encoded signal.

In the spin echo formation, the prephasing gradient is also used to spatially encode the spins. However, after this gradient is applied, at $t = TE/2$, a refocusing RF pulse is used to perform a ‘pancake flip’ or mirror the spins across the axis along which the RF pulse is applied (Figure 3.6d). The spins continue to accrue phase due to the frequency encoding gradient until they are, once again, in phase, and form the spin echo at $t = TE$ (Figure 3.6e). In spin echo pulse sequences, the prephaser and frequency encoding gradients have the same polarity and are placed on opposite sides of the RF refocusing pulse.

The advantage of using spin echoes is that the transverse signal decay depends solely on T_2 relaxation, as the phase from all other field inhomogeneities is eliminated at $t = TE$. In gradient echoes, the local field inhomogeneities cannot be reversed, and the echo is dependent on T_2^* . On the other hand, gradient echoes can be acquired with lower effective TE, as the need for an additional RF refocusing pulse is eliminated. Another important consideration in MR imaging is the time between successive RF excitation pulses, called the *repetition time* (TR). In general, TR is selected to maximize longitudinal relaxation. Gradient echo pulse sequences can use RF excitation pulses with very low flip angles (57), thus allowing lower TRs than spin echo pulse sequences. Many fast MRI pulse sequences adopt gradient-echo techniques.

The two imaging parameters, TR and TE, are important because, in most pulse sequences, these parameters can be optimized to yield the best image contrast for a particular application. Image contrast will depend upon tissue T_1 relaxation values if TR and TE are selected to be much smaller than the estimated T_1 and T_2 parameters for that tissue. Likewise, selecting longer TR and TE can emphasize the tissue T_2 relaxation parameters. Image intensity can also depend upon the spin density, or *proton density*, of the tissue, if the TR is much longer than T_1 , and the TE is shorter than T_2 . T_1 - and T_2 -weighted images

provide a deeper understanding of healthy and diseased tissue, and are frequently used to aid in clinical diagnoses.

The following sub-sections discuss the basic spin and gradient echo pulse sequences. The pulse sequences are shown as timing diagrams with the RF excitation and refocusing (RF), slice selection (SS), phase (PE) and frequency (RO) encoding on separate axes.

3.4.1 Spin Echo Pulse Sequence

Radiofrequency spin echo or spin echo (SE) sequences employ 180° refocusing RF pulses to form the echo signal. The advantage in using spin echoes is that the 180° pulse refocuses off-resonance effects arising from main magnetic field inhomogeneity and local susceptibility variations. Spin echo pulse sequences can also be used to generate images with T_1 -, T_2 -, or proton-density weighting, by varying the TR and TE times. The pulse sequence for a spin echo image is shown in Figure 3.7.

A spin echo sequence employs a 90° RF excitation pulse and a 180° RF refocusing pulse, to form one spin echo after every TR. The slice profiles corresponding to the RF excitation and refocusing pulses are closely matched. Crushers are placed on either side of the slice refocusing gradient lobe, as discussed in Section 3.2.1. The sequence also consists of phase and frequency encoding steps. The number of repetitions depends upon the number of phase encoding steps, and the number of excitations per phase encoding step. The total time taken for a SE pulse sequence can therefore be very long. An alternative is to use the fast spin echo pulse sequence to acquire more than one phase encoding step within each TR (see Section 4.2). A spoiler gradient is applied on the slice selection axis and played out at the end of each acquisition.

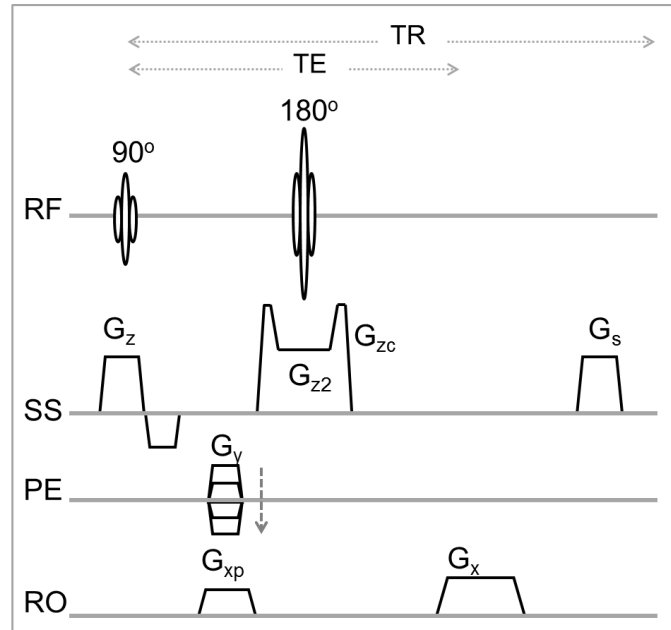


Figure 3.7: The basic spin echo pulse sequence, showing the order of the radiofrequency pulses (RF), slice selection (SS), phase (PE) and frequency (RO) encoding as a function of time.

Variations on the basic spin echo pulse sequence can increase its efficiency and applications. Usually, the slice-rephasing lobe is combined with the left crusher gradient lobe, to decrease the minimum TE. The spoiler can be placed on any axis, and is often bridged with the readout gradient on the frequency encoding axis. Partial k -space data can be acquired by adjusting the areas of the prephaser and the readout gradients (see Section 3.2.2). The pulse sequence can be adapted to perform chemically or spatially selective saturation or magnetization transfer imaging. Phase rewinders can be added at the end of the pulse sequence to further reduce artifacts from the remaining transverse magnetization. This is discussed further in Section 3.4.2. Reduced field-of-view images can be acquired without artifacts by placing the slice selection gradient (G_z) on the phase encoding axis (34). A technique similar to this has been introduced in this thesis to address the cusp artifact seen in non-axial fast spin echo images (see Chapter 5).

3.4.2 Gradient Echo Pulse Sequence

Gradient echo (GE) is a class of pulse sequences that is usually used for fast scanning (55,56). As discussed earlier, in these sequences, gradient reversal on the frequency encoding axis is used to form the echo, instead of the 180° RF refocusing pulse. Gradient echoes are also called *gradient recalled echoes* (GRE) or *field echoes*. Figure 3.8 shows the pulse sequence diagram for gradient echo image acquisition.

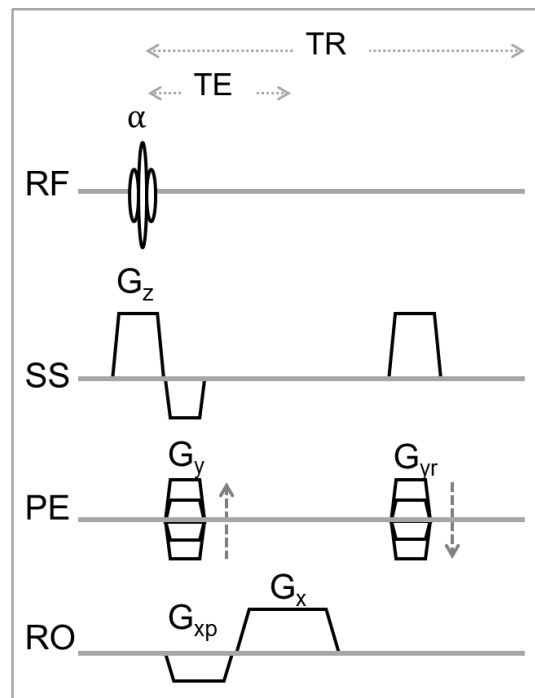


Figure 3.8: The basic gradient echo pulse sequence

With a primary exception of the RF refocusing pulse, many of the components of the GRE pulse sequence are similar to those discussed for the SE pulse sequence. Slice selection is followed by the

rewinder gradient, which can be played concurrently with the phase encoding gradient and the prephasing gradient on the readout axis, to decrease TE. As gradient echoes are T_2^* -weighted, GRE sequences are designed such that the TE is as small as possible. Figure 3.8 shows a full-echo acquisition, where the area of the prephasing gradient is designed to be half of that of the readout gradient. However, GRE sequences very frequently employ partial Fourier acquisitions. The Fast Low-Angle SHot (FLASH) technique that was proposed along with gradient-echo based imaging (57) utilizes small flip angles (α) for RF excitation. The flip angle is selected to cause enough transverse magnetization to be detected while keeping intact most of the longitudinal magnetization for use during the following repetitions. This allows the TR to be maintained at a minimum, increasing the data acquisition speed.

Although the primary reason for using GRE is data acquisition efficiency, their unique contrast make them preferred in several applications. For example, local susceptibility variations can be mapped using the T_2^* contrast provided by GRE, an attribute exploited in susceptibility weighted imaging (58) and iron detection (59). GRE sequences can provide images with hyperintense blood signal, and are used in phase-contrast imaging (60). The data acquisition speed of GRE allows us to image anatomy in real time, e.g., in cardiac imaging (23,61).

3.5 Pulse Sequences for Specific Applications

Spin and gradient echo pulse sequences form the basis of nearly all the image acquisition protocols in MRI. The basic pulse sequences shown in Figure 3.7 and Figure 3.8 can be modified to allow the acquisition of images of optimal quality for specific applications, as discussed in Sections 3.4.1 and 3.4.2. These applications include perfusion imaging using contrast agents or spin labeling techniques, phase-contrast MR angiography, separation of fat and water in tissues, and imaging of molecular diffusion, among others.

Perfusion is the process by which nutritive arterial blood is supplied to the tissues, and metabolic by-products are drained into the veins. MRI can be used to measure perfusion of blood in tissues through

the use of contrast agents such as gadolinium chelates. This technique is called *dynamic susceptibility contrast* (DSC) imaging (62), and it involves the intravenous injection of a contrast agent, immediately followed by the acquisition of a series of images of each slice, at several times. The pixel intensity plotted as a function of time reveals the flow of the contrast agent through the imaged tissue. Quantitative measures such as the perfusion rate, blood flow, and mean transit time (63,64) can be determined from these images.

Another perfusion technique, called *arterial spin labeling* (ASL) (65), relies on the creation of an endogenous tracer. The magnetization from flowing arterial blood is labeled at an upstream location, through the use of a saturation or inversion pulse. The signal intensity changes as the arterial blood perfuses into the tissue. Once again, a series of images is acquired rapidly to observe the change in signal intensity over time. Arterial spin labeling can yield the same perfusion-related measures as DSC imaging.

Either SE- or GRE-based pulse sequences can be used for perfusion imaging applications. However, perfusion imaging using ASL essentially relies on the signal difference between control and tagged images; which is only 1~2%. This small signal change necessitates the use of signal averaging to increase SNR. EPI (66), FSE (27), and spiral (49) imaging are therefore preferred in ASL-based perfusion sequences.

Phase contrast (PC) MRI (60) is also used to measure moving magnetization, especially the flow of blood within vessels, and cerebrospinal fluid. To image flow using MRI, a bipolar flow-encoding gradient is used to produce a phase which is directly proportional to the velocity of flowing fluids. The axis of the bipolar gradient determines the direction of flow sensitivity, and the first moment of the gradients determines the highest velocity encoding possible. Typically, PC pulse sequences are formed by adding the bipolar gradients to a FLASH pulse sequence (67), although sequence variations, such as steady-state free precession (SSFP) (68), spiral imaging (49), and EPI (66) have also been proposed.

A widely used application of MRI is in the suppression and/or the separation of the protons in fat from the protons in water. The lipid signal can interfere with correct diagnosis of hyperintense lesions in T_2 -weighted imaging, and can obscure contrast-enhanced lesions in T_1 -weighted imaging, due to its low

T_1 relaxation time. This problem is significant because fat signals contribute to severe chemical shift artifacts (69) in the body. Techniques for lipid suppression include the use of saturation pulses or inversion pulses (70) applied before the RF excitation, or the use of spatial-spectral RF excitation (71) pulses, such as those used in EPI.

Some applications require the isolation of the lipid signal from the water signal, commonly referred to as fat-water separation. This is usually performed by considering a two-component model where the spectral peaks come from either fat or water. As these two signals have a distinct phase difference (~ 3.5 ppm), it is possible to use the principle of echo shifting (72) to acquire two images: one with fat and water signals in-phase, and another with the two signals out-of-phase. A pure fat and pure water image can be generated from the in-phase and out-of-phase images. This technique, called Dixon's method (73), was proposed in 1984, and has spawned many variations (74,75).

The following section will focus on diffusion imaging. Ever since the effects of molecular diffusion in NMR were first discussed in 1954 (76), the application of MRI to diffusion imaging has increased several-fold. Many pulse sequences have been developed specifically for diffusion imaging (77,78). Diffusion imaging is discussed in more detail in the following section.

3.5.1 *Diffusion Imaging*

Molecular diffusion is characterized by Brownian motion, and is therefore essentially random. In a system without restrictions, the radius of diffusion, r , is a function of the diffusing time, t , and the diffusion coefficient, D , and is given by:

$$r = \sqrt{2Dt} \quad [3.10]$$

In biological tissues, diffusion of water molecules is affected by the presence of macromolecules, cell membranes, organelles, fibers, and other cellular structures. When a magnetic field gradient is applied to diffusing water molecules, the result is a phase dispersion of the transverse magnetization (76), which results in signal loss. This means that MRI, which employs linear magnetic field gradients, can be used to

measure diffusion in tissues. Many pulse sequences can be rendered sensitive to diffusion, although RF spin echo-based sequences, such as fast spin echo and spin-echo echo planar imaging, are commonly used. Of these, EPI-based sequences are preferred due to their high data acquisition speed and inherent insensitivity to motion.

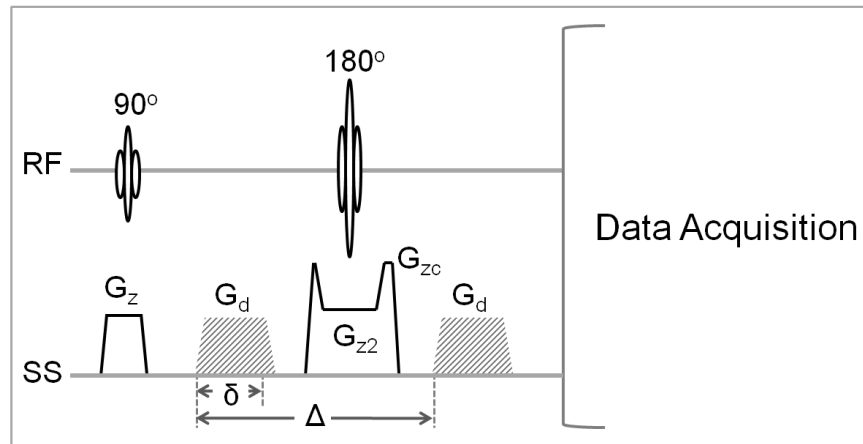


Figure 3.9: Diffusion preparation using a spin echo-based pulse sequence. The left (G_{dl}) and right (G_{dr}) diffusion-weighting gradient lobes are shaded in gray. In this example they are placed on the slice selection axis, although they may be placed on any one or a combination of gradient axes.

To increase the sensitivity of MRI to diffusion in tissues, all diffusion imaging pulse sequences contain a pair of *diffusion-weighting gradients* (79), as seen in Figure 3.9. The amplitude (G_d), duration (δ), and separation (Δ) of the gradient lobes can be adjusted to produce the desired diffusion weighting, given by a constant called the *b*-value. The gradients work in a manner analogous to phase-contrast imaging. Assuming a homogeneous magnetic field, the first gradient lobe, which is played before the RF refocusing pulse, produces a finite phase dispersion in the magnetization. The second lobe is played after

the refocusing pulse, and rephases the stationary spins. The result of local molecular diffusion is signal loss, given by:

$$S = S_0 e^{-bD} \quad [3.11]$$

where S and S_0 are the signal intensities with and without diffusion. This model of diffusion signal attenuation is commonly referred to as the monoexponential model. The faster the diffusion, the larger is the phase dispersion due to the diffusion-weighting gradient, and the greater is the net signal attenuation. The diffusion signal loss is only detected along the direction of the diffusion-weighting gradient.

Diffusion imaging comprises several specific techniques that include diffusion-weighted imaging (DWI), diffusion tensor imaging (DTI) (80), diffusion spectroscopic imaging (DSI) (81), high angular resolution diffusion imaging (HARDI) (82,83), and q-space imaging (84). Diffusion-weighted imaging employs a single b-value to produce an image. A set of images formed with two or more b-values can be used to create a quantitative map of diffusion coefficients across the image. In diffusion tensor imaging, apart from a non-diffusion-weighted image, a set of at least six diffusion-weighted images is acquired with the same b-values, with each image having diffusion sensitivity in a different direction. These images can yield scalar and vector maps with information about the nature, size, and orientation of the structures within the imaging region of interest. Diffusion images are usually analyzed using a variety of diffusion models of signal decay due to molecular diffusion, e.g., mono-exponential (given by Eq. 3.11), bi-exponential (85), statistical (86), fractional-order calculus-based (87), and kurtosis (88) models. These models have been reviewed in a recent article (89).

3.6 Gradient Non-Ideal Behavior

The behavior of the gradient system depends upon its characteristics. Although the system is designed to exhibit linear behavior over a large field of view, with amplitudes and slew rates as high as possible, in practice, it may not be possible to achieve the target values due to imperfections within the gradient system. These imperfections result in some form of spatial distortion or a loss of signal intensity,

which can be seen on the image. This section briefly discusses the most commonly seen non-ideal behaviors in the gradient system, with an emphasis on their effect on the obtained image.

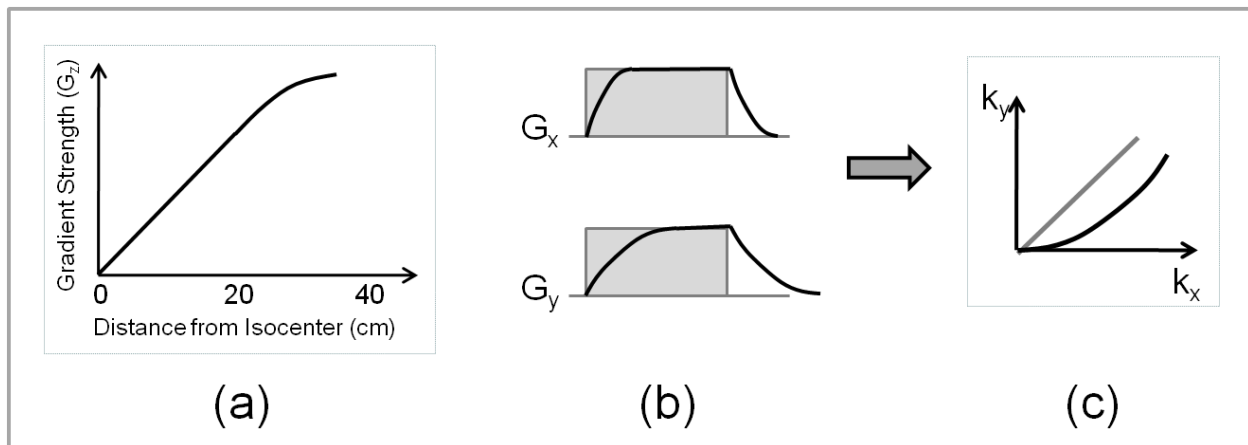


Figure 3.10: Cartoon illustrating non-ideal gradient behavior, such as (a) gradient non-linearity due to which the effective amplitude of the gradient decreases away from the isocenter, (b) eddy currents generated in the gradient coils which decrease the slew rate of the gradient fields (shown in black), and (c) the effect of gradient anisotropy on the k -space trajectory (in black) compared to the target trajectory (in gray).

3.6.1 Non-Linearity

It is not possible to generate a perfectly linear gradient in space. Although gradient coil designers can design gradient fields that are linear over a limited volume or region, it is necessary to quantify and account for distortions or other artifacts associated with non-linear gradients. Gradient coils are usually designed to adhere to some measure of linearity, speed and image quality-based specifications. The region over which the linearity is evaluated is called the linearity volume. Gradient non-linearity is defined as the

difference between the actual value attained by the gradient field and its ideal value, normalized to the ideal value. In most whole-body MRI machines, the maximum gradient deviations are within 5% of the gradient ideal (also called desired or target) value (37).

Within the plane of the image, the non-linear gradient away from the isocenter is well-known as a cause of geometric distortion of the image, and can be corrected (90). Gradient non-linearity can also cause the slice thickness to alter along the slice plane, resulting in a slice of uneven thickness or a ‘potato-chip’ shape.

3.6.2 Eddy Currents

Eddy currents are induced as a result of the changing magnetic fields within the gradient coils, and usually result in lower effective gradient amplitude along with a phase lag. EPI sequences are particularly sensitive to eddy currents due to the strong and rapidly switching acquisition gradients. As an example, the x -component of the eddy current field can be modeled by the following Taylor series expansion (47):

$$B_e(\vec{x}, t) = b_0(t) + \vec{x} \cdot \vec{g}(t) + \dots \quad [3.12]$$

where $b_0(t)$ is the spatially constant B_0 eddy current, which can be modeled by:

$$b_0(t) = -\frac{dG_x}{dt} \otimes e_{0x}(t) - \frac{dG_y}{dt} \otimes e_{0y}(t) - \frac{dG_z}{dt} \otimes e_{0z}(t) \quad [3.13]$$

The vector $\vec{g}(t) = [g_x(t) \ g_y(t) \ g_z(t)]$ is the *linear eddy current* which has linear spatial variation. Each of the three components (in x , y , and z) can be modeled by an equation such as:

$$g_x(t) = -\frac{dG_x}{dt} \otimes e_{xx}(t) - \frac{dG_y}{dt} \otimes e_{xy}(t) - \frac{dG_z}{dt} \otimes e_{xz}(t) \quad [3.14]$$

Each of the impulse response functions is a sum of decaying exponentials. The first term in Eq. 3.14 is the eddy current produced in the x -gradient field by the applied x -gradient, and is called the direct linear term.

The second and third terms are the eddy currents produced in the x -gradient field by the applied y - and z -gradients, respectively. These terms are therefore called cross terms and they are usually much smaller than the direct terms. Eq. 3.12 also indicates the existence of higher-order eddy current terms; however, these terms are not significant and are therefore not usually considered. In general, the B_0 eddy current causes a constant, spatially independent phase across the image, whereas the linear eddy current results in a linear phase variation on the image. Most eddy-current-related problems with image quality result from one of these two terms.

Eddy current reduction is a many-step process. First, most MRI scanners employ gradient coil shielding, where a shield or outer coil is designed to produce a field that opposes the field from the primary gradient coils (91). This compensates for most of the eddy currents, and leaves only an uncompensated fringe field. This uncompensated field can be further reduced using pre-emphasis (92), where the gradient waveform is reshaped such that the pre-emphasis distortion cancels the subsequent distortion produced by the eddy currents. Finally, eddy currents can be further corrected using additional acquisition (93) or post-processing steps (94), depending upon the application. For example, constant and linear phase can be corrected using a statistical approach to estimate these errors directly from the imaging data (95).

3.6.3 Anisotropy between Gradient Axes

The behavior of each gradient channel of an MRI scanner can be approximated by an impulse response relating the actual gradient to the gradient specified by the pulse sequence. Ideally, this impulse response is a Dirac delta function, but this is never achieved in practice due to one or more system imperfections, including unmatched gradient delays or eddy current behaviors between gradient axes. The gradient system is said to be anisotropic if the impulse responses on any two or more of the three channels differ from each other (32). Gradient anisotropy is only seen on oblique images, where more than one

physical gradient axis is used for excitation and image acquisition. Usually, the x - and y -axes exhibit very similar gradient behavior.

The MR image quality can be considerably degraded due to gradient anisotropy. The nature and amount of distortion are related to the imaging parameters describing both, RF excitation and image acquisition. The most commonly seen impact of gradient anisotropy is in the distortion of the k -space trajectory, especially in echo-train or gradient-echo-based pulse sequences, such as EPI and spiral imaging. For a given application or pulse sequence, the degree of anisotropy can be estimated from information about gradient delays and eddy current characteristics, obtained from the MRI scanner. This information can be used to correct for anisotropy, in real-time or after image acquisition (96).

4 Pulse Sequences for Rapid Imaging

This chapter describes pulse sequences for rapid imaging. High data acquisition speeds are usually obtained through echo-train pulse sequences, where multiple spin or gradient echoes are produced after every RF excitation pulse. Several phase-encoded lines of k -space can then be acquired after every RF excitation pulse, decreasing the total acquisition time by a factor equivalent to the number of echoes following each RF excitation pulse. These encoding schemes allow the available transverse magnetization (M_{xy}) to be used in an efficient manner. In this chapter, three fast acquisition strategies are discussed: echo planar imaging (EPI), fast spin echo (FSE), and periodically rotated overlapping parallel lines with enhanced reconstruction (PROPELLER).

4.1 Echo Planar Imaging

Echo planar imaging (Figure 4.1a) was the first rapid imaging technique proposed for MRI, in 1977 (66,97). Using EPI, it is possible to acquire an entire image or set of images within a single TR, i.e., ~ 4 sec. This makes EPI a very attractive pulse sequence to observe dynamic physiological processes such as neural activation (98), diffusion (99) and perfusion (100). Although it remains one of the most efficient imaging pulse sequences till date, it is also technically very demanding, which, until recently, limited its use in both clinical and research studies. EPI is also prone to severe artifacts arising from eddy currents, B_0 inhomogeneity, and magnetic susceptibility arising from air-fluid interfaces.

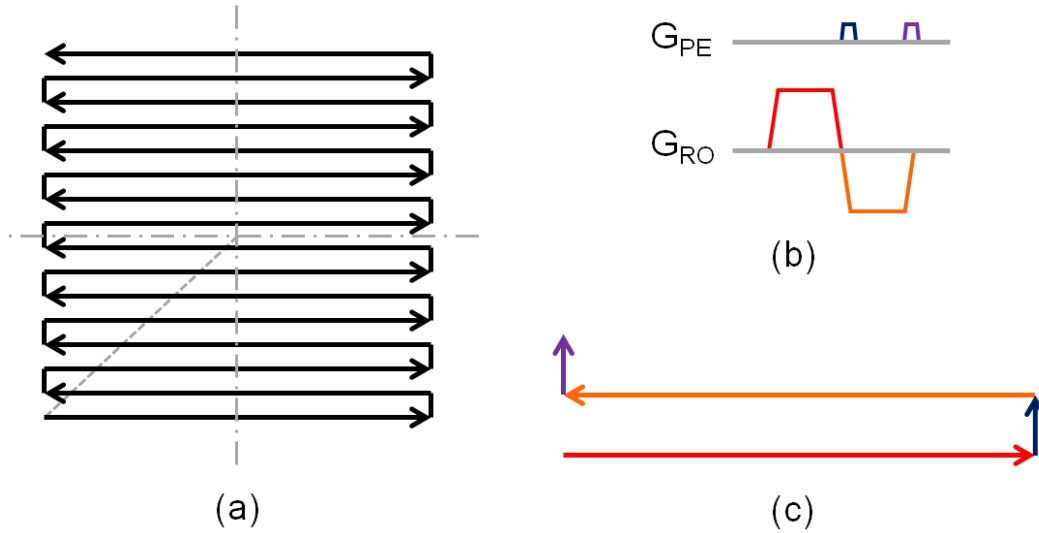


Figure 4.1: (a) shows the EPI sampling strategy. The bipolar readout gradient (G_{RO}) and the phase-encoding blip gradient (G_{PE}) pulses that are used to sample this trajectory are color-coded in (b) to illustrate the corresponding sections of the trajectory in (c).

4.1.1 Pulse Sequence

The basis of fast acquisition in EPI lies in the sampling of several lines of k -space within a single TR (Figure 4.1a). This is accomplished by employing a rapidly-switching bipolar readout gradient (Figure 4.1b) so as to acquire echoes in a back-and-forth scheme. Each echo is individually phase-encoded, so that a set of k -space lines can be acquired within one shot. The number of phase-encoded lines acquired within a shot is called the echo train length (ETL), and determines the scan time. Figure 4.2 and Figure 4.4, respectively, show the pulse sequences for gradient- and spin-echo-based EPI acquisition schemes.

An EPI pulse sequence starts with a selective slice-excitation RF pulse. As EPI typically suffers from a large chemical shift artifact due to lipid signals, schemes for lipid suppression are generally used. Figure 4.2 and Figure 4.4 show spatial-spectral (SPSP) excitation pulses which are designed to exclude

lipid signals. Alternatively, a saturation pulse can be applied to reduce signals from lipid. As EPI typically uses long TRs, the flip angle of the excitation pulse is set to 90° , to increase the signal.

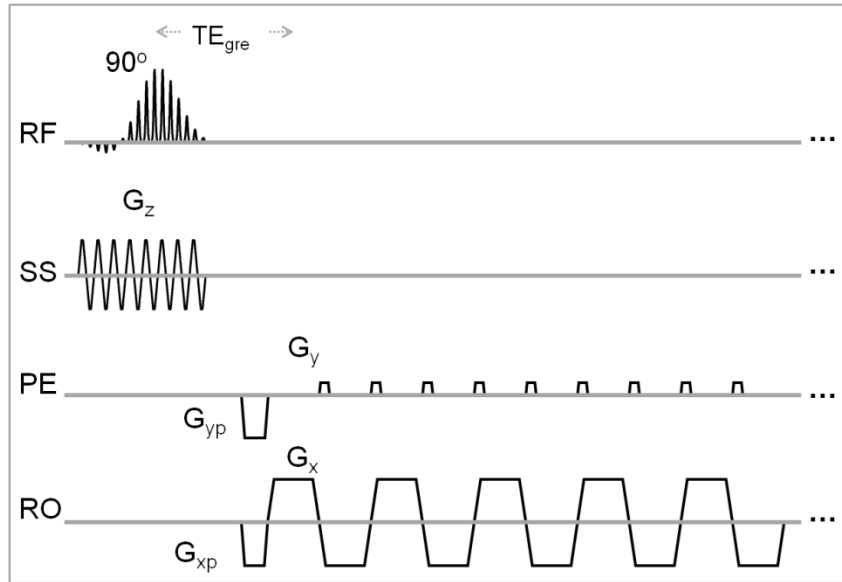


Figure 4.2: A gradient-echo echo planar imaging pulse sequence. (The SPSP excitation pulse structure was provided by Yi Sui at the University of Illinois at Chicago.)

The FID signal produced by the selective excitation pulse in EPI is manipulated by a series of readout (G_x) and phase-encoding (G_y) gradients. The readout gradient is preceded by a prephaser (G_{xp}), designed with an area half of that of G_x to coincide the center of the echo with the center of the acquisition window. The phase-encoding gradient (G_y) is also preceded by a prephasing lobe (G_{yp}) that sets the position of the initial line of k -space in the phase-encoding direction. The area, A_{G_y} , of each phase-encoding “blip” gradient (G_y) is given by the equation:

$$A_{Gy} = \frac{2\pi}{\gamma FOV_y} \quad [4.1]$$

where $FOV_y = 1/\Delta k_y$ is the FOV along the phase-encoding direction. The blips can be designed as trapezoidal or triangular waveforms. Although Figure 4.2 and Figure 4.4 show the phase-encoding acquisition order going from the lower to the upper half of k -space, other acquisition schemes are also used, based on the application.

In the gradient echo EPI scheme (Figure 4.2), the signal decay along the echo train (S_n) is given by the equation:

$$S_n = S_0 e^{-TE_n/T_2^*} \quad [4.2]$$

where S_0 is the signal at the isocenter of the RF pulse and TE_n is the echo time of the n th echo in the echo train. The effective TE (TE_{eff}), different from the gradient-echo TE shown in Figure 4.2, is given by the TE of the phase-encoding line corresponding to the center of k -space, and can be altered by changing the amplitude of G_{yp} . Figure 4.3a shows a single-shot GRE-EPI image.

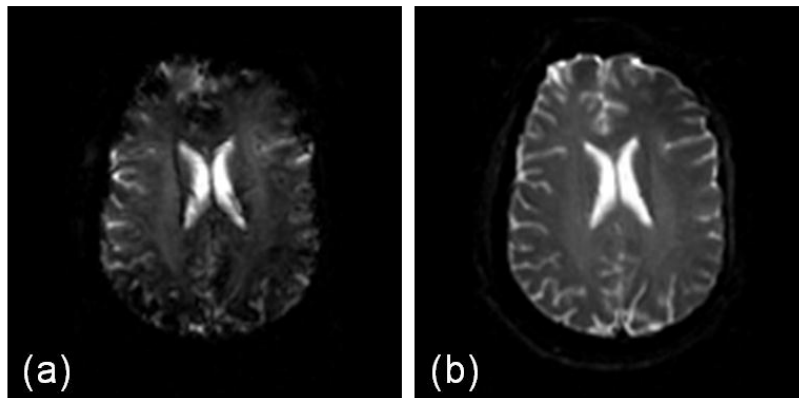


Figure 4.3: Single-shot (a) Gradient-echo EPI, and (b) Spin-echo EPI images. TR/TE = 3000/90 ms, acquisition matrix 128×128.

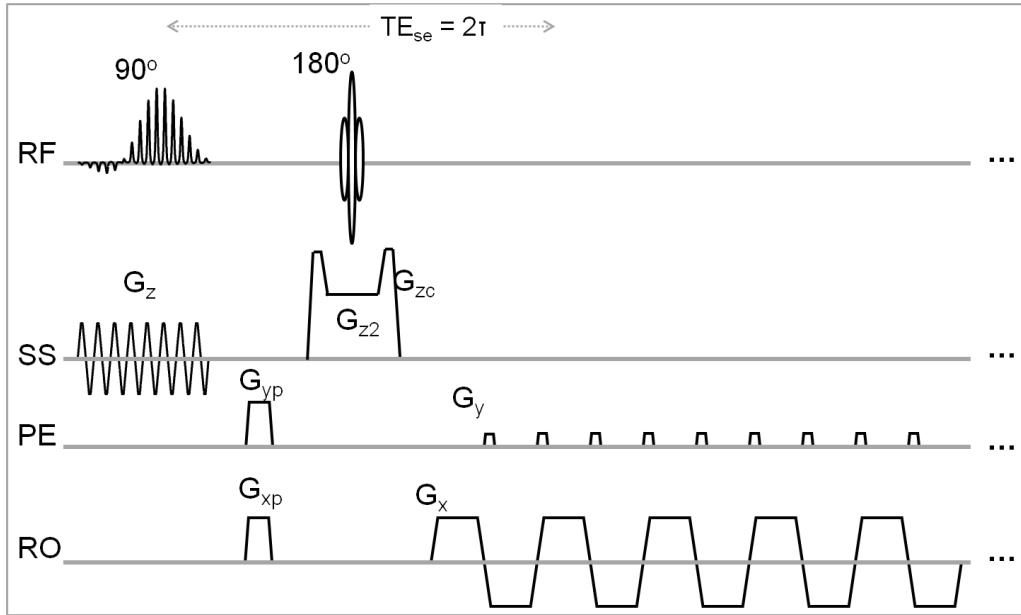


Figure 4.4: A spin-echo echo planar imaging pulse sequence. (The SPSP excitation pulse structure was provided by Yi Sui at the University of Illinois at Chicago.)

The RF excitation pulse in the spin-echo EPI pulse sequence is designed in the same manner as that in the GRE-EPI pulse sequence. In the spin-echo EPI pulse sequence (Figure 4.4), the RF excitation pulse is followed by a slice-selective refocusing RF pulse. The SE-EPI pulse sequence employs the same data acquisition scheme as the gradient-echo EPI sequence. The data is acquired under the envelope of the spin echo, with the highest spin echo amplitude at $TE_{se} = 2\tau$. The remaining echoes on either side of the spin echo in the train decay with a time constant of T_2^* . The effective TE, however, is still defined in the same manner as in GRE-EPI, as the TE associated with the echo acquired at the center. Crushers are used on either side of the refocusing RF pulse to eliminate unwanted signals. Spoiler gradients are employed in both, SE- and GRE-EPI pulse sequences, to dephase any transverse magnetization left over after the acquisition. A representative single-shot SE-EPI image is shown in Figure 4.3b. Figure 4.3a and Figure 4.3b are acquired with the same parameters, and the distinction between the spin- and gradient-echo

versions of EPI can be seen clearly in the signal dropouts due to increased B_0 inhomogeneity and susceptibility artifacts in the gradient-echo image.

4.1.2 Features

The two most significant characteristics that determine the speed of EPI acquisitions are the echo train length (ETL) and spacing between echoes (ESP). In order to sample as much of the trajectory as possible, ETL is usually kept very large. On the other hand, longer prescribed ETLs lead to longer durations of the readout gradient, which, in turn can lead to severe image distortion and exacerbate the Nyquist ghost. One way to reduce the total duration of the readout gradient is to reduce the time spacing between consecutive echoes (ESP). Assuming that the echo spacing, t_{esp} , is determined by the total time duration of the readout gradient alone:

$$t_{esp} = T_{acq} + 2T_{ramp} = \frac{2\pi n_x}{\gamma FOV_x G_x} + 2 \frac{G_x}{SR} \quad [4.3]$$

where T_{acq} is the duration of the flat-top of the readout gradient, n_x is the number of points in the readout, FOV_x is the FOV in the readout direction, and G_x is the amplitude of the readout gradient. T_{ramp} is the ramp time, which depends upon the slew rate, SR . From Eq. 4.3, t_{esp} can be decreased by increasing the readout bandwidth, and proportionally increasing the amplitude of the readout gradient. An alternative approach is to sample on the ramps of the readout gradient. Doing so can reduce the pulse-width of the constant part of the gradient lobe, thus reducing the ESP.

Reconstructing images acquired using EPI involves a few additional steps. First, the k -space lines in EPI are acquired in a back-and-forth scheme. During reconstruction, alternate echoes need to be flipped so that the direction of all the lines is the same. EPI data usually contains artifacts arising from inconsistent eddy currents, B_0 inhomogeneity and chemical shift, which are usually corrected for after row-flipping (101). In order to decrease the echo spacing, partial acquisition schemes are usually

employed in EPI. Reconstruction can therefore involve several additional steps, or algorithms such as homodyne processing (102).

4.1.3 Artifacts in EPI

One of the most common artifacts in EPI is the Nyquist ghost. Nyquist ghosts predominantly arise from eddy currents in large, rapidly switching readout gradients, which result in lower effective readout gradient amplitudes along with a phase lag. EPI sequences are particularly sensitive to eddy currents due to the strong and rapidly switching acquisition gradients, which lead to constant (Figure 4.5a) and linear (Figure 4.5b) phase Nyquist ghosts, as discussed in Section 3.6.2. Techniques that have been developed to reduce constant and linear phase ghosts commonly employ reference scans (93). A reference scan is the EPI data acquired by disabling the phase-encoding gradient during data acquisition. The phase error, or inconsistencies, can be obtained by taking the Fourier transform of each line of the reference scan, and comparing the obtained phases with a reference. As this data is acquired empirically, it most accurately reflects the phase errors for a particular imaging protocol, and can be subsequently used for phase correction.

A larger inconsistency in the eddy-current behavior among gradient axes (see Section 3.6.3 on Gradient Anisotropy) (32) can lead to k -space shifts in the phase encoding direction of the blade, resulting in the formation of the oblique Nyquist ghost (ONG, Figure 4.5c) (33,103). The strength of the ONG in EPI depends upon the relative anisotropy between gradient axes, and the plane rotation angle. The most common scheme for its correction involves using the gradient delays on each physical gradient axis to design compensatory blips along the phase-encoding direction (96,104).

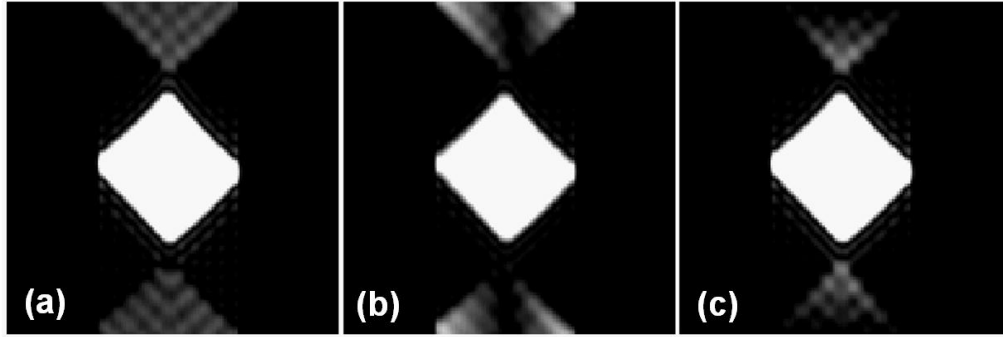


Figure 4.5: Phantom images illustrating (a) constant and (b) linear phase Nyquist ghosts in the readout direction, and (c) oblique Nyquist ghost in the phase-encoding direction, in EPI.

Unlike other sampling schemes, chemical shift artifacts manifest in the phase-encoding direction in EPI. This is because the readout bandwidth in EPI is large enough to suppress this artifact in the readout direction. As an example, a phase-encoding bandwidth of 2 kHz can lead to a chemical shift of 27 pixels along the phase-encoding direction for a FOV of 24 cm at 3.0 T. Lipid suppression schemes, such as the SPSP excitation pulse, are always used in EPI to reduce the artifact.

EPI also suffers from severe image distortion from off-resonance effects such as B_0 field inhomogeneity, magnetic susceptibility variations, and concomitant fields. In general, the refocusing RF pulse employed in an SE-EPI pulse sequence renders SE-EPI images less vulnerable to artifacts arising from inhomogeneity and susceptibility than GRE-EPI, as illustrated by Figure 4.3. However, SE-EPI is also less sensitive to blood oxygenation level dependent (BOLD) contrast, which is susceptibility-dependent.

4.1.4 Variations

In EPI, an entire set of k -space lines may be acquired within the time it takes for most of the transverse relaxation to occur, i.e., ~ 100 ms. This acquisition scheme is called single-shot EPI (ssEPI, Figure 4.6a). In ssEPI, high data acquisition speeds are maintained by the use of high-amplitude gradient fields and large readout gradient bandwidths. It is, therefore, the sequence of choice for dynamic imaging experiments such as functional MRI, diffusion, and perfusion imaging. Its short acquisition time means that ssEPI is generally insensitive to subject motion. However, ssEPI is limited to lower spatial resolutions (i.e., upto 128×128), and is prone to artifacts due to its demanding hardware requirements.

Distortions in EPI occur as a result of long readout durations. To reduce the readout duration in ssEPI, sampling schemes in ssEPI usually acquire only a part of k -space. The rest of k -space can be filled by using the property of conjugate symmetry. Partial filling of k -space is done in two ways: (a) by acquiring only a part of each echo, called partial-echo or partial k_x acquisition, and (b) by acquiring only a fraction of the echoes (or phase-encoding steps), called partial k_y acquisition. In either scheme, only half of the k -space is needed to reconstruct the image. In practice, however, a little more than half of k -space is usually acquired, to correct for phase shifts from inhomogeneity, eddy currents and other sources of phase error. In ssEPI, a partial k_y sampling scheme is usually used.

In multi-shot EPI (msEPI, Figure 4.6b), a part of the k -space is acquired after every excitation pulse. This acquisition scheme is possible by varying the amplitudes of the readout and/or phase-encoding prephasing gradients, so that a different set of lines in k -space can be acquired in each shot. This scheme allows the acquisition of higher-resolution images. As it is possible to maintain shorter ETLs, the demands on the imaging hardware are lower, allowing the acquisition of images with higher SNR, and less geometric distortion and blurring, and lesser ghosting. The bandwidth along the phase-encoding direction is also increased when compared with ssEPI, reducing phase-encoding artifacts. Multi-shot EPI, however, may require additional correction due to subject motion between shots.

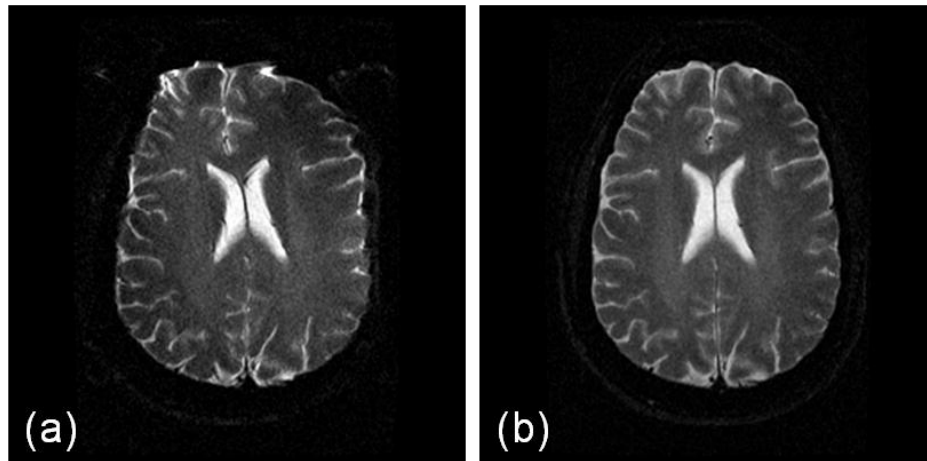


Figure 4.6: (a) Single-shot spin echo EPI, and (b) multi-shot spin-echo EPI images. $TR/TE = 3000/90$ ms, acquisition matrix 256×256 .

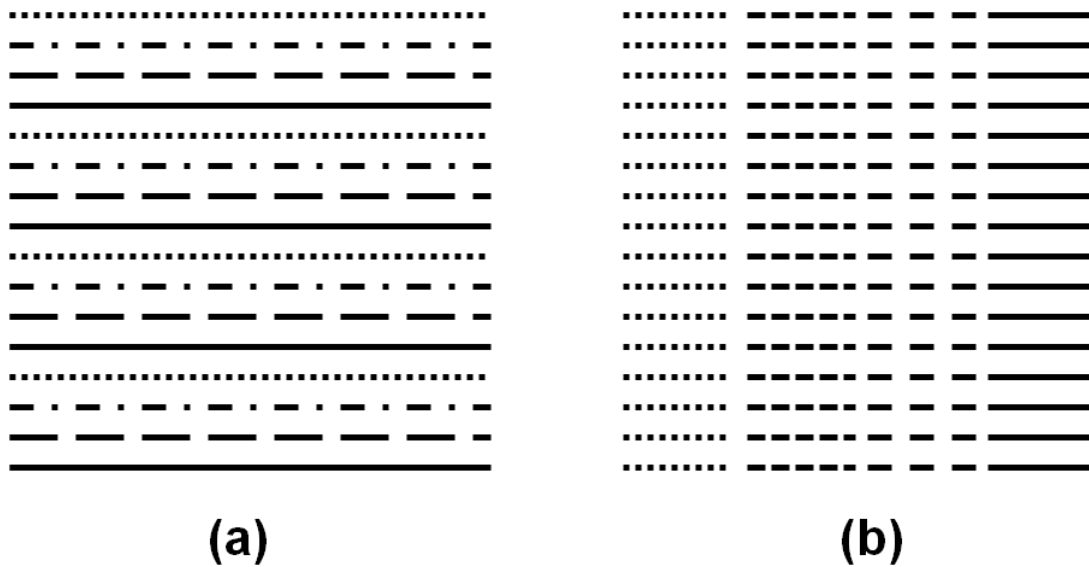


Figure 4.7: (a) Interleaved, and (b) readout-segmented acquisition schemes for multi-shot EPI.

Several acquisition schemes for msEPI have been proposed. In the most commonly used scheme, the amplitude of the phase-encoding prephasing gradient is changed for every shot. Along with this, the spacing between phase-encoding lines is also increased. Together, these changes allow an interleaved set of lines to be acquired within each shot (Figure 4.7a). An alternative acquisition scheme to reduce the ESP has been proposed (105). In this scheme, a series of consecutive segments (or blinds) is acquired within each shot, where each blind acquires a part of the readout for a full set of phase-encoding steps (Figure 4.7b). This sampling technique, called readout-segmented EPI (RS-EPI, (106)), has demonstrated a reduction in image distortion and chemical shift artifact.

Other variations possible in EPI acquisition are skip-echo EPI (107), where every other k -space line is acquired, thus not using every alternate readout gradient lobe for data acquisition. This obviates the need for phase correction, and results in an image free of Nyquist ghosts. Another sampling scheme is circular EPI (108), where the k -space trajectory is designed to sample data within a circle. The circular sampling results in higher data acquisition efficiency, as, in general, the corners of k -space do not provide sufficient information. A true three-dimensional version of the EPI pulse sequence, called echo volumar imaging (EVI), was also proposed (29,109). This scheme is rarely used for anatomical imaging due to its very low spatial resolution, but has acquired momentum in functional imaging applications (110).

4.2 Fast Spin Echo

A conventional spin echo (SE) sequence can take as long as fifteen minutes to acquire a set of images. In the standard SE, the data is usually acquired for only a small fraction of the time that the transverse magnetization can be recorded, resulting in low data acquisition efficiency. The *fast spin echo* (FSE) pulse sequence (which is one of the several trade names of the *rapid acquisition with relaxation enhancement*, or RARE, pulse sequence) (27) was based on the echo imaging principle used in EPI, and created as a rapid sampling alternative to the SE sequence to avoid the need to correct for the artifacts produced in the EPI sequence. In the FSE sequence, an RF excitation pulse is followed by a train of RF

refocusing pulses to produce a set of spin echoes. Each spin echo is distinctively phase-encoded so that several lines of k -space can be acquired within a single excitation. The number of primary spin echoes produced after every excitation pulse is referred to as the *echo train length* (ETL). This factor also determines the scan time reduction in an FSE pulse sequence. Figure 4.8 shows a single excitation of an FSE pulse sequence.

4.2.1 Pulse Sequence

The FSE pulse sequence consists of a slice-selective 90° RF excitation pulse, followed by a train of selective 180° RF refocusing pulses (76), which are commonly shifted by a phase of 90° away from the excitation pulse (111) to satisfy the Carr-Purcell-Meiboom-Gill (CPMG) conditions. The CPMG conditions are meant to improve the robustness of the pulse sequence to B_1 -field non-uniformities and to ensure that coherent echoes are formed at the same temporal position within the pulse sequence. These conditions dictate that:

1. The time interval between successive 180° pulses (2τ) is twice that between the 90° and the first 180° pulse (τ). The refocusing RF pulses must be 90° out of phase with respect to the excitation RF pulse.
2. The phase accumulated by a spin isochromat between any two consecutive RF refocusing pulses must be equal.

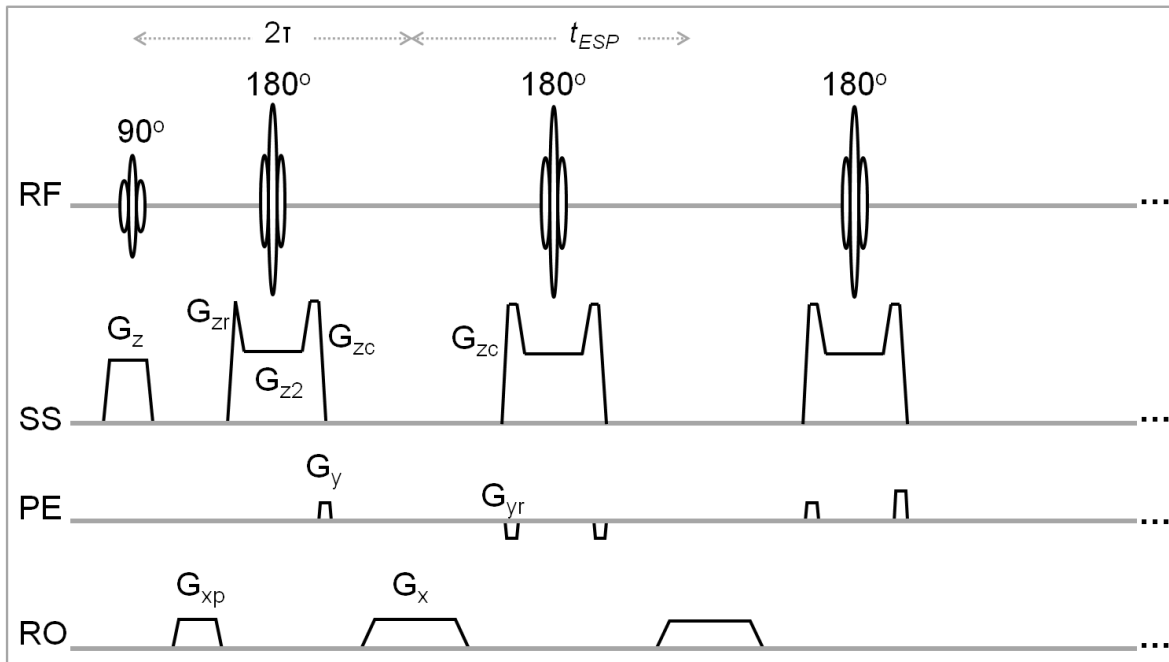


Figure 4.8: A fast spin echo pulse sequence.

The RF excitation and refocusing pulses are slice-selective, i.e., they are played concurrently with slice-selection gradients (G_z). A slice rephasing gradient lobe is played after the slice selection gradient. Crushers (G_{zc}) are placed around the slice refocusing gradient (G_z) to select the desired signals and exclude the unwanted signals. To satisfy the second CPMG condition, the left and right crusher lobes are designed to have the same area. The first left crusher lobe (G_{zr}) is usually combined with the slice selection rephasing gradient lobe.

In the FSE sequence, several phase-encoded lines of k -space are acquired within each shot (or TR). Identical readout gradients (G_x) are played exactly halfway between each set of consecutive RF refocusing pulses. The time between any pair of consecutive readouts, called the echo spacing time (ESP or t_{esp}), depends upon factors such as the readout bandwidth and determines the speed of data acquisition. The first readout gradient (G_{xp}) is preceded by a prephaser. This gradient lobe is designed with area half

of that of the readout gradient such that the center of the echo is acquired at the center of the readout. For subsequent readout gradients (after the first), the second half of the previous readout gradient is also the prephaser for the next.

Identically designed phase encoding (G_y) and phase rewinding ($G_{yr} = -G_y$) gradient lobes are used before and after each readout gradient. The areas of these gradient lobes vary along the ETL, and are calculated to acquire ETL number of distinctive phase-encoding steps within each shot. The order in which the phase-encoding steps are acquired within the echo train determines the image contrast. At the end of each excitation, a spoiler gradient (G_{yk} , not shown in Figure 4.8) is usually applied in the phase-encoding direction to dephase the residual transverse magnetization.

The total scan duration (T_{scan}) for an FSE pulse sequence with N_{PE} phase encoding steps is:

$$T_{scan} = TR \times NEX \times \frac{N_{PE}}{ETL} \quad [4.4]$$

This equation holds for an interleaved multislice dataset in the case when all the slice locations can be acquired within the same TR. If the number of slice locations exceeds this limit, the scan duration increases by a factor of N_{acq} , which is the number of excitations required to acquire the entire set of slices, such that:

$$T_{scan} = TR \times NEX \times \frac{N_{PE}}{ETL} \times N_{acq} \quad [4.5]$$

4.2.2 Features

Each echo in the echo train of an FSE pulse sequence is acquired at a different echo time; this means each echo within the echo train has different T_2 -weighting (27). The T_2 -weighting function for a given tissue, $W(k_y)$ can be given by:

$$W(k_y) = e^{-n(k_y) \times t_{esp} / T_2} \quad [4.6]$$

where n denotes the position of the echo in the echo train, and t_{esp} is the echo spacing time. As the echoes acquired at the center of k -space contribute the most to the image contrast, the effective TE (TE_{eff}) in an FSE pulse sequence is given by the TE of the echo acquired at the center of k -space. The contrast of the image can therefore be altered by changing the order of acquisition of the echoes in the center of k -space. T_1 - and proton-density weighted images (Figure 4.9a and b, respectively) are produced by using the early echoes in the echo train to sample the center of k -space. Similarly, to acquire T_2 -weighted images (Figure 4.9c), the late echoes in the echo train are used to sample the center of k -space. Dual-echo FSE sequences have been developed to acquire proton-density and T_2 -weighted images simultaneously. The proton-density-weighted images are generally acquired using the earlier echoes in the train, and the T_2 -weighted images use the late echoes.

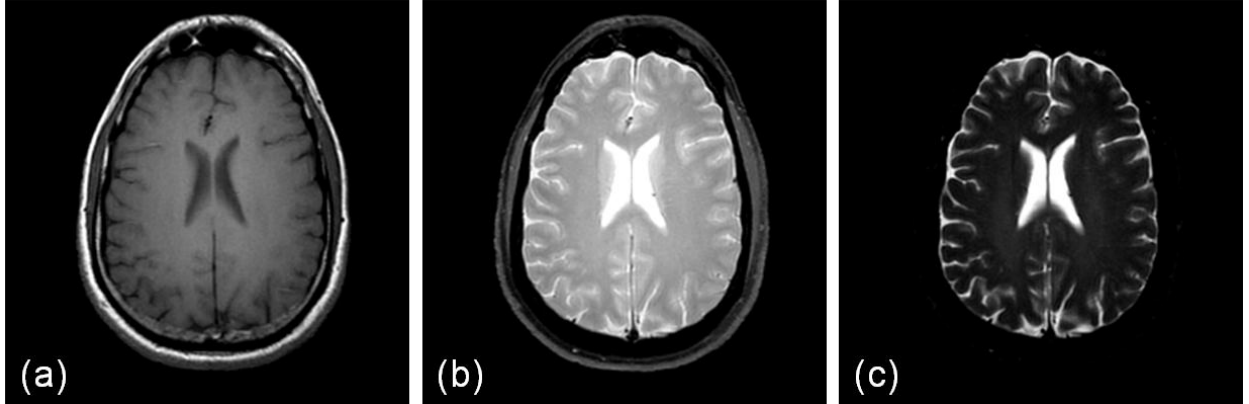


Figure 4.9: Fast spin echo images acquired on a healthy volunteer with (a) T_1 -weighting (TR/TE = 600/12 ms), (b) proton-density-weighting (TR/TE = 3000/20 ms), and (c) T_2 -weighting (TR/TE = 3000/100 ms).

The FSE pulse sequence was introduced not only as a rapid alternative to the conventional SE pulse sequence, but also as a way of reducing the off-resonance and inhomogeneity artifacts arising from fast gradient echo pulse sequences, such as EPI. This is possible if the CPMG conditions are satisfied, in which case the temporal positions of the primary and stimulated echoes are the same (τ , 3τ , 5τ , etc.), and the spin echoes are perfectly refocused, leading to high-quality images.

4.2.3 *Artifacts in FSE*

Images acquired using FSE typically suffer from artifacts that manifest along the phase-encoding direction, e.g., blurring, the formation of ghosts, and the cusp artifact. As discussed in Section 4.2.2, each acquired echo is increasingly T_2 -weighted based on its position within the echo train. The k -space data in FSE is therefore equivalent to the k -space signal in SE weighted by a function that approximates the T_2 signal decay for that echo position. Upon Fourier transformation, this weighting function assumes a Lorentzian shape that effectively results in the blurring along the phase encoding direction. The blurring increases as the T_2 -weighting increases; for example, as ETL or spacing between echoes (ESP) increases, or in tissues with shorter transverse relaxation time.

Due to the refocusing pulses in FSE, the phase accrued from off-resonance and inhomogeneity errors is cancelled at the peak of each echo. However, phase may also accrue from sources such as eddy currents, group delays, and concomitant gradients. If the phase is consistent across k -space, the magnitude images do not show any errors. Inconsistent phase errors cause ghost artifacts to appear on the magnitude images. Several correction techniques have been proposed to reduce these artifacts. Constant and linear phase errors can be reduced by using a non-phase-encoded reference-scan based-method (112), similar to that proposed for EPI. Concomitant gradient fields can be corrected for by redesigning aspects of the FSE pulse sequence (113), such as the crusher gradients surrounding the slice refocusing pulses.

Another artifact that occasionally arises in the phase encoding direction is the cusp artifact. This artifact was first observed on sagittal and coronal FSE images of the spine and knee (31), and is one

among a class of aliasing artifacts and occurs as a result of gradient non-linearity away from the isocenter (26). Although the cusp artifact is not unique to FSE, it has a distinctive feather-like signal smearing pattern in the phase-encoding direction of an FSE image that can overlap with and obscure important image features. With technological advances in magnet design, this artifact is expected to be seen more frequently and prominently on non-axial FSE images. The FSE cusp artifact has been discussed in detail in Chapter 5, and its reduction is one of the objectives of this thesis.

4.2.4 Variations

One practical consideration in the design of an FSE protocol is the management of the specific absorption rate (SAR). SAR is the measure of the RF power (in Watts) dissipated in a mass of tissue (in kilograms). SAR is a particularly important factor in FSE due to the use of multiple RF pulses within the echo train. SAR can be managed in a number of ways. In FSE, the number of slices per acquisition is usually limited to limit the RF pulses in the echo train. Variable-rate RF pulses are also used to decrease the B_1 -field amplitude (114).

Several other modules can be appended to FSE pulse sequences. Signal from fluid can be suppressed by using a FLAIR (FLuid Attenuated Inversion Recovery) pulse sequence, where an inversion pulse is placed before the RF excitation pulse. Spatial or chemical saturation pulses can also be applied to suppress lipid signals, which appear especially bright in FSE. An alternate technique (115) uses the GRASE (GRAdient and Spin Echoes) pulse sequence framework to acquire off-resonance FSE images which are then used for both, lipid suppression and susceptibility weighting.

4.3 Gradient and Spin Echo (GRASE)

The gradient and spin echo (116,117) pulse sequence is a combination of EPI and FSE. This pulse sequence employs a train of refocusing pulses, such as in FSE, combined with a series of alternating

readout gradients following each refocusing pulse, such as in EPI. The result is a pulse sequence that overcomes some of the challenges in the use of both, EPI and FSE, while inheriting some of their limitations.

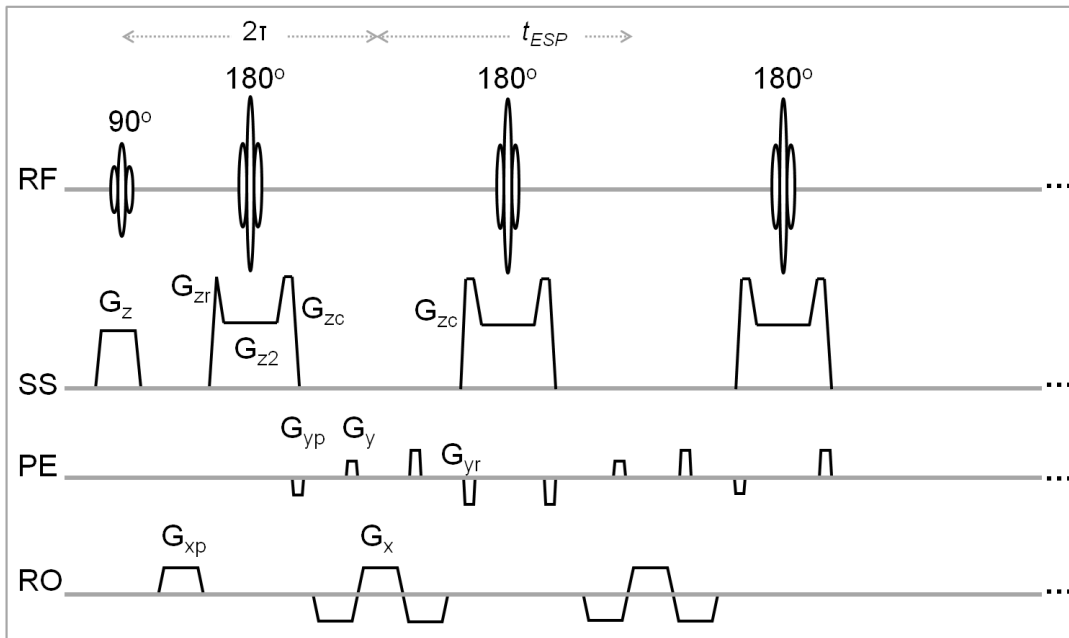


Figure 4.10: The gradient and spin echo (GRASE) pulse sequence where three gradient echoes follow every spin echo.

In GRASE (Figure 4.10), each RF refocusing pulse is followed by a train of individually phase-encoded gradient echoes. The phase-encoding gradient is completely rewound before the next refocusing pulse, to satisfy the CPMG conditions, similar to FSE. This additional step allows data acquisition to increase by a factor of N_{gre} (which is the number of gradient echoes following each RF refocusing pulse) compared with a standard FSE sequence. As a result, fewer refocusing pulses are required, which results

in a reduction in SAR as well as in the total scan time. The reduction in the total acquisition time is reflected in the equation:

$$T_{scan} = TR \times NEX \times \frac{N_{PE}}{ETL \times N_{gre}} \times N_{acq} \quad [4.7]$$

Alternately, images with higher resolution can be acquired in the same time given by Eq. 4.7. The number of gradient echoes (N_{gre}) is carefully selected to obtain the required T_2^* contrast. Most commonly, three or five phase-encoded gradient echoes are acquired after every refocusing pulse.

The GRASE pulse sequence results in a lower lipid signal than that seen in FSE. For an image with the same number of excitations, the total acquisition time per excitation is lower than in FSE, and results in less blurring than the use of the FSE sequence. Due to these advantages over FSE, a single-shot GRASE pulse sequence has also been proposed (118). These advantages are offset by some of the limitations of GRASE. For example, the use of bipolar gradient echoes results in inhomogeneity and off-resonance effects, although these are far less conspicuous in GRASE than in EPI. GRASE is also sensitive to eddy currents due to acquisition along both, positive and negative readouts, and usually requires phase-correction, similar to that used in EPI. GRASE can be highly sensitive to the T_2^* weighting, which results in amplitude modulation and ghosting artifacts on the image. For this reason, the phase-encoding view order is very important.

4.4 The PROPELLER Sampling Technique

The PROPELLER (Periodically Rotated Overlapping Parallel Lines with Enhanced Reconstruction) method (30) is a novel MRI sampling technique that traverses k -space using a series of rectangular strips rotated about the k -space origin (Figure 4.11). It offers excellent robustness against motion due to the inherent navigator information collected by means of sampling the central region of k -space for every strip. Additionally it provides a number of other advantages, most notably, reduced

sensitivity to magnetic susceptibility variations and warping effect, in the presence of magnetic field inhomogeneities and eddy currents. PROPELLER is increasingly being used in a number of applications, primary among them being high-resolution diffusion-weighted (DW) imaging (77).

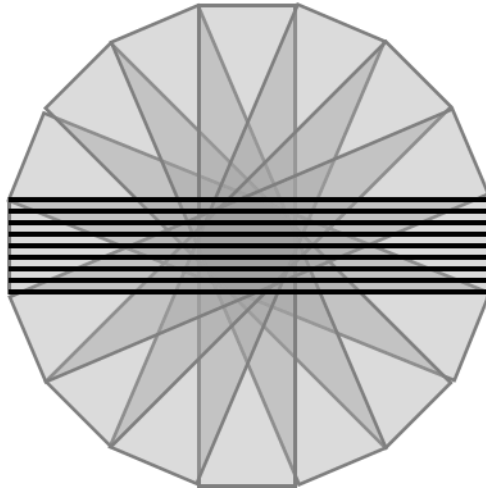


Figure 4.11: The PROPELLER sampling scheme consists of several blades (rectangles) arranged uniformly around the center of k -space. Each blade is sampled using one of several Cartesian sampling schemes.

Data acquisition in PROPELLER is performed by the collection of N strips, or blades, of data, each containing L lines of k -space. The L lines are spaced by $\Delta k_y = 1/L_y$, where L_y is the phase-encoding FOV for that blade. In general, all blades have the same Δk_y , which leads to a perfectly circularly sampled k -space. However, k -space can also be sampled non-isotropically using this method (30,119). The minimum number of blades, N , can be calculated using the relationship:

$$N = \frac{M}{L} \frac{\pi}{2} \quad [4.8]$$

where M is the desired acquisition matrix along the readout direction. This relationship assumes no overlap between blades at the outermost point. The angle between two consecutive blades (also called the blade angle) is given by $\theta = \pi/N$. The number of blades can be increased from the minimum by using a smaller blade angle.

In order to acquire a single blade within a single excitation, any multi-echo Cartesian-based data acquisition strategy can be used. The PROPELLER technique was initially implemented using FSE (77). Since then, EPI-based PROPELLER sampling schemes (120,121) have been developed. A few GRASE-based PROPELLER schemes (122-124) have also been proposed.

4.4.1 Data Acquisition

The PROPELLER technique was initially implemented in a multi-shot fast spin echo (FSE) pulse sequence. In this implementation, each spin echo in the FSE echo train is used to acquire one line of k -space data, thus the entire echo train produces a blade or strip consisting of L parallel k -space lines (Figure 4.12a) where L equals the echo train length (ETL). As seen in Figure 4.13a, PROPELLER based on FSE inherits many desirable properties of FSE, especially the excellent robustness against off-resonance effects due to the use of multiple refocusing RF pulses (Figure 4.13a). The disadvantage is that, compared to other fast imaging techniques, PROPELLER based on FSE may not offer the data acquisition efficiency in some demanding applications. For example, in diffusion tensor imaging (DTI), diffusion-weighted images with multiple gradient directions must be acquired at the same slice location. FSE-based PROPELLER sequences can take more than 10 minutes to acquire a DTI dataset, which limit their use in clinical applications. Additionally, the large number of radiofrequency (RF) pulses in FSE-PROPELLER can escalate the specific absorption rate (SAR), especially at higher magnetic fields such as 3.0 T.

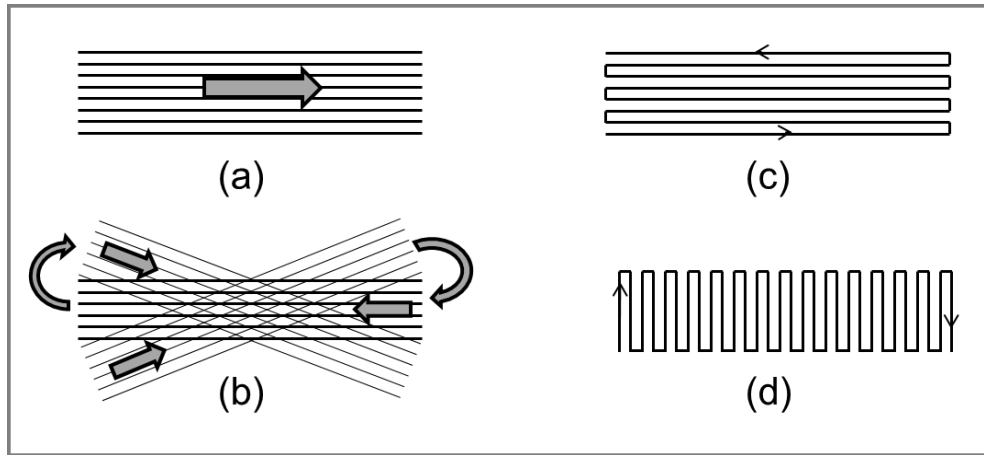


Figure 4.12: Possible sampling schemes in PROPELLER imaging include (a) FSE, (b) GRASE-based Steer-PROP, (c) long-axis EPI-sampling, and (d) short-axis EPI-sampling.

To address the issues with efficiency and SAR in FSE-PROPELLER, GRASE-based PROPELLER schemes have been proposed. GRASE is a technique that can combine the merits of FSE and EPI. In turboprop proposed by Pipe (122), a PROPELLER sequence is implemented in GRASE. Central to turboprop is the addition of a short EPI train inside every spin echo of the FSE echo train by replacing the conventional frequency encoding gradient with an oscillating one as well as collecting multiple phase-encoded lines of data. This serves to widen the blade, thus improving the robustness of motion correction because a large amount of redundant data at the k -space central region can be used for evaluating data consistency. Additionally turboprop facilitates the acquisition of the entire k -space using lesser number of blades and hence improves data acquisition efficiency without escalating SAR problems. Like other GRASE sequences, the view order is very important and contributes to the mitigation of phase errors arising from off-resonance effects within a blade. Turboprop also faces the challenges of phase correction within each blade. The inter-blade phase errors can become complicated because of the two inter-tangled sources: phase errors arising from violation of the CPMG conditions (i.e., FSE-type phase error) and phase errors due to the EPI-type k -space traversal. A split-blade approach has been proposed to

reduce errors arising from violations of the CPMG conditions (125). In turboprop acquired with the split-blade approach, the odd and even spin echoes are distributed along two orthogonal blades.

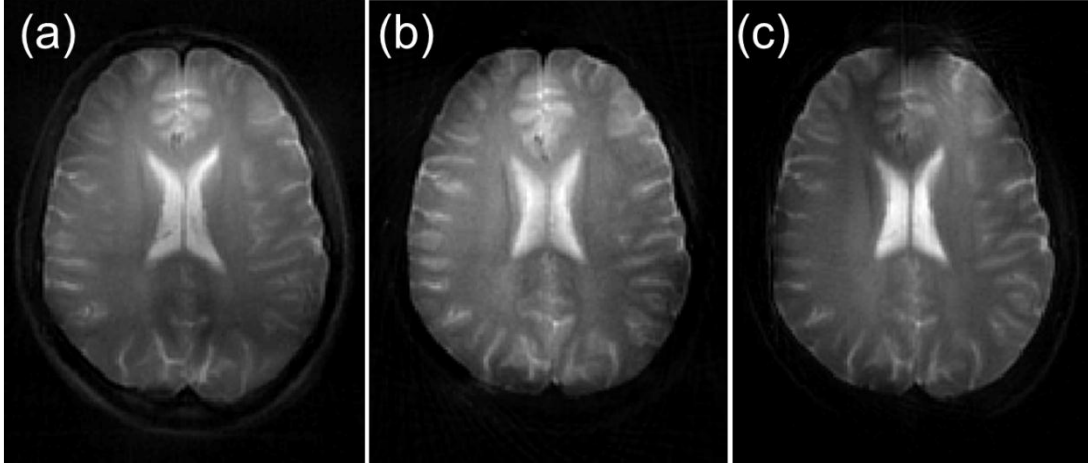


Figure 4.13: Volunteer images using (a) FSE-based PROPELLER, (b) Steer-PROP, a GRASE-based PROPELLER pulse sequence, and (c) long-axis EPI PROPELLER pulse sequences. TR/TE = 4000/72 ms for all sequences.

In an effort to address the phase errors arising from the bipolar gradients, an alternative GRASE-based PROPELLER technique, called *Steer-PROP*, has been developed and implemented (123). Unlike turboprop, Steer-PROP traverses N_{gre} k -space blades in one shot where N_{gre} is the number of gradient echoes following every RF refocusing pulse in the GRASE echo train. A series of blip pulses employed between the gradient echoes distribute the N_{gre} gradient echoes to N_{gre} different blades. This acquisition strategy is illustrated in Figure 4.12b. This technique allows for increased data acquisition efficiency up to a factor of N_{gre} without increasing the blade width. For example, the number of excitations required to

acquire a full set of blades in FSE-PROPELLER is the same as the number of blades, given by N in Eq. 4.8. The number of shots (N_{TR}) can further be reduced by a factor corresponding to N_{gre} , in Steer-PROP:

$$N_{TR} = \frac{N}{N_{gre}} \quad [4.9]$$

Steer-PROP also provides an effective way to address the problems with phase errors inherent to GRASE-based PROPELLER. The three types of phase errors in GRASE-based PROPELLER are: (a) inter-shot phase errors arising primarily from motion, (b) inter-blade phase errors among the blades acquired within a shot (i.e., the phase errors among the gradient echoes), and (c) intra-blade phase errors among the k -space lines sampled by different spin echoes in the GRASE echo train. The first two types of phase errors can be corrected using the redundant data at the center of k -space. This correction is discussed in further detail in Section 4.4.2. The third type of phase error arises from violation of the CPMG conditions in FSE, and can be corrected by acquiring two non-phase-encoded lines for every blade to calculate the phase inconsistencies (see Section 4.2.3). Images acquired using Steer-PROP have demonstrated very similar quality compared with FSE-PROPELLER. A sample image is shown in Figure 4.13b.

Recently, EPI-based PROPELLER sequences, including long-axis PROPELLER (LAP, (120)) and short-axis PROPELLER (SAP, (121)), were introduced as strategies to significantly improve the data acquisition speed in PROPELLER and make it capable of recording dynamic processes (such as diffusion and perfusion) efficiently. In EPI-based PROPELLER, each blade is acquired using a single-shot spin-echo echo-planar sampling strategy (Figure 4.12c,d). EPI allows faster sampling than FSE, resulting in the acquisition of wider blades which provide better motion correction due to the increase in the redundancy between blades. EPI-based PROPELLER dispenses with the multiple RF refocusing pulses used in FSE- and GRASE-based PROPELLER so that the SAR is no longer a concern. However, large readout gradients in EPI can lead to larger geometric distortion and signal decay due to off-resonance effects and magnetic susceptibility. Also, the lower phase-encoding bandwidth in a traditional EPI-based

sequence can accentuate chemical shift artifacts. To mitigate these effects, the readout gradient can be applied along the short axis of the PROPELLER blade. This acquisition technique is referred to as short-axis PROPELLER (SAP-EPI, (121)), seen in Figure 4.12d, as against the long-axis PROPELLER (LAP-EPI, (120)), in Figure 4.12c, where the readout direction is along the long axis of the PROPELLER blade. Although some of the drawbacks can be addressed by using SAP instead of LAP, phase errors arising from eddy currents are a significant challenge in the application of both LAP and SAP. These phase errors are modulated by the blade angle and typically require blade-specific correction, which is usually performed using reference scans or phase-maps, increasing the total scan duration substantially. Figure 4.13c shows a phase-corrected volunteer image acquired using LAP-EPI.

4.4.2 Reconstruction and Artifact Correction

PROPELLER is a non-Cartesian data acquisition technique especially designed to correct motion artifacts before reconstruction using the Fourier transform. In general, the reconstruction pipeline consists of intra-blade phase correction, inter-blade motion correction, blade-correlation-based weighting to reduce the contribution from blades that are likely corrupted by in-plane and through-plane motion, and pixel density weighting and gridding to a standard Cartesian grid. As in Cartesian image reconstruction, the FFT is usually the final step. Figure 4.14 shows a flowchart of the various operations involved in PROPELLER image reconstruction.

Phase correction is usually performed on a blade-by-blade basis. Data acquired using FSE are corrected for phase errors that arise due to violations of the CPMG conditions (*see* Section 4.2.3), whereas the data acquired using EPI are corrected for phase errors occurring as a result of eddy currents that arise from the rapid switching of gradient fields (*see* Section 4.1.3). As this step is performed on a per-blade basis, several schemes formulated for Cartesian imaging strategies can be used.

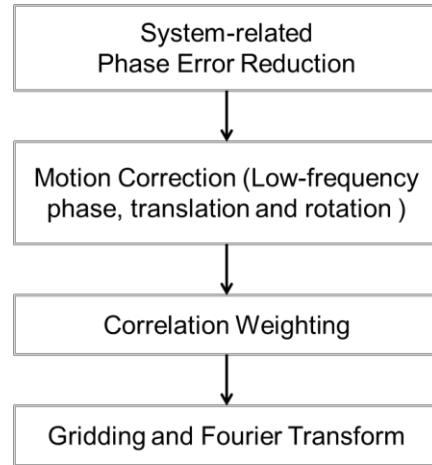


Figure 4.14: Flowchart illustrating the steps involved in PROPELLER image reconstruction.

Next, the data are corrected for translational and rotational motion. This consists of two primary steps, the first of which is to ascertain that the point of rotation of the blade corresponds with the center of k -space. The displacement of the k -space center results in a spatially varying phase on the image. Additionally, low-frequency motion can also cause a phase variation which can be different for the image reconstructed from each blade. These phase variations are removed by using a triangular window on each blade. The resultant k -space is the difference of the original and the windowed k -space.

Rotation and translation between blades are corrected by using the oversampled center of k -space. To estimate rotation, the magnitudes of the central oversampled region from each blade are averaged to form a reference dataset. Next, the central region of each blade is rotated by a series of angles and gridded onto the locations in the averaged grid. The correlation between each blade and the average is measured as a function of the rotation angle, and fit to a polynomial to estimate the angle of rotation for each blade. In a similar manner, translation between blades can be estimated by using the complex-valued averages of the overlapping central region instead of the magnitudes. The translation of each blade from the average

is obtained by finding the maximum value of the convolution between each blade and the averaged blade. Translation and rotation values are subtracted from each blade.

Once the rotation and translation effects are removed, the central data are averaged again for correlation-weighting. A correlation measure is calculated for each blade with this new average, and the contribution of the blade is weighted in proportion to its correlation with the average. A threshold is set for inclusion of each blade, based on its correlation weight. This step allows one to reject blades that may have large through-plane motion or uncorrected in-plane motion. This is the final step before the corrected data are gridded to Cartesian co-ordinates before applying the Fourier transform. Gridding in PROPELLER also comprises a density-weighting operation by which the data points away from the center are emphasized (47). This removes the weighting at the center of k -space which has a much higher signal due to oversampling.

5 Reduction of the FSE Cusp Artifact

This chapter discusses the cusp artifact seen on non-axial fast spin echo images, especially on the human spine and knee. The theory behind the formation of the cusp artifact is explained, along with a discussion of the imaging parameters that influence either the appearance or the strength of the artifact. Next, a pulse-sequence design-based approach to reduce the cusp artifact is proposed and validated on phantoms and human volunteers. This chapter then concludes with a discussion of this technique, including a comparison of the slice tilting approach with other commercial approaches and a summary of possible challenges to the application of the technique. Finally, the slice tilting approach is proposed for a different application, i.e., the reduction of aliasing artifacts in diffusion-weighted PROPELLER imaging.

5.1 Background

The magnetic field gradient in an MRI scanner is linear only within a limited region near the magnet isocenter. Beyond this region, virtually all gradient systems display non-linear spatial characteristics, particularly at or near the edge of the magnet (see Section 3.6.1). This non-ideal condition is exacerbated by a rapid change of main magnetic field (B_0) towards the end of the magnet bore. As a result, the overall magnetic field produced by the combination of the gradient field and the B_0 field has a rather complicated spatial dependence. At a region away from the isocenter (marked as C in Figure 5.1), the overall magnetic field experienced by spins can be equal to the net magnetic field at or near the magnet isocenter (marked as ‘A’ in Figure 5.1) (26). The region away from the isocenter, which has been called the “gradient null” (126), is typically outside the imaging volume of interest. If a radiofrequency (RF) coil (or a coil element in a phased array) receives signals from that region, the signal will carry the same (or similar) frequency as the signal near the isocenter, leading to an aliasing artifact in the image. In

a fast spin echo (FSE) sequence, the aliasing artifact manifests itself as a series of spots, a band, or a ‘feather-like’ artifact at or near the center of the field of view (FOV) along the phase-encoding direction. The artifact is often observed on sagittal or coronal planes in spine and knee scans (31), interfering with image interpretation. In the literature, this artifact has been called cusp artifact, anefact, fold-over artifact, feather artifact, peripheral signal artifact, along with other names (26,31,126). Although the cusp artifact does not appear in exactly the same form (i.e., ‘C’-shaped as shown in (31)) under specific conditions, the mechanism of the artifact formation remains substantially the same. In this work, we refer to this artifact as “cusp artifact”, this term having been used in several other publications (26,31,126).

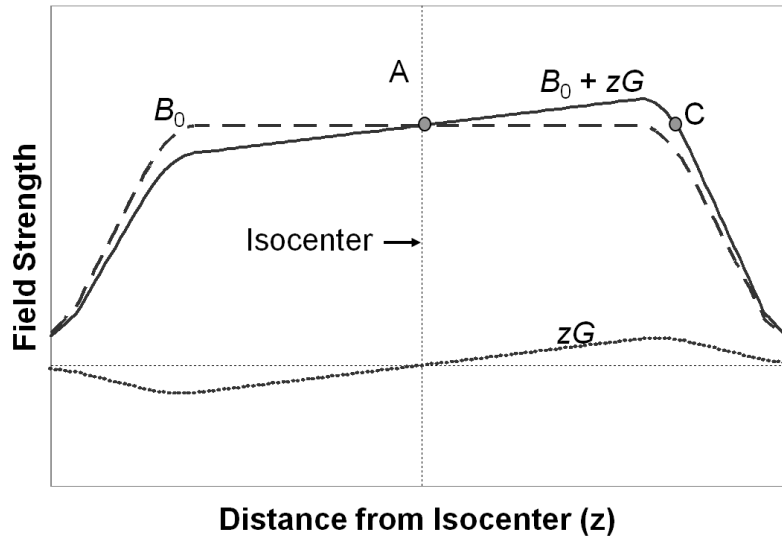


Figure 5.1: Profile of $B_{tot} = B_0 + z.G$ through the bore of the scanner (adapted from (26)).

5.1.1 *Currently Available Reduction Techniques*

One of the commercial techniques to reduce the artifact relies on adaptive phased array (APA) coils (127). Individual elements of a phased array coil can be chosen automatically by an algorithm that determines the proper coil elements based on user-specified FOV while rejecting the signals from coil elements at or near the artifact-prone regions. Although the current literature does not contain a comprehensive evaluation of this approach, it has been observed effective under specific conditions (e.g., imaging with limited FOV). To implement APA, substantial modification of the RF receiving electronics and a signal selection algorithm are needed. To avoid the hardware changes, signal-processing techniques based on parallel imaging (e.g., sensitivity encoding or SENSE (16)) have been used to reduce the FSE cusp artifact (126,128). These techniques estimate a B_1 sensitivity matrix by utilizing two separate coils, one placed at the magnet isocenter and the other (smaller in size) at or near the artifact-producing region. The non-aliased signal within the FOV can be recovered using a parallel-imaging reconstruction algorithm (16). This approach assumes that the approximate location of the artifact-producing region is known, and requires a calibration procedure to estimate the sensitivity matrix for each RF coil. The assumption and the need for calibration can impose a problem in practical implementation. Another approach entails the use of a metal foil (also known as “metal skirt” or “RF blanket”) over the artifact-producing region to dephase the magnetization leading to the artifacts. This technique can be effective when the RF blanket is positioned exactly at the location of the artifact source. However, it can impose a safety concern due to the possibility of increased local heating (129). The safety concern can become prohibitive in a SAR-intensive sequence, such as FSE, particularly at high magnetic fields (e.g., 3.0 T). Other techniques of limited scope have also been proposed (31,126). In the following section, a novel pulse-sequence design-based approach is proposed to reduce the cusp artifact. This approach is similar in theory to the inner-volume imaging techniques (34) that have been proposed specifically for smaller FOVs (130). This proposed technique requires no modifications of the hardware or in the reconstruction process, and the virtually artifact-free images can be viewed immediately after acquisition.

5.2 Theory

To reduce the FSE cusp artifact, our approach is to modify the FSE pulse sequence so that the slice selected by the RF excitation pulse is slightly tilted with respect to the slice selected by the subsequent RF refocusing pulses. With this modification, peripheral magnetization that would cause the cusp artifact will not experience both RF excitation and refocusing pulses, leading to artifact reduction or even elimination.

5.2.1 Pulse Sequence Design

This method was implemented in a commercial FSE pulse sequence (GE Healthcare, Waukesha, WI) in which a small “slice-selection” gradient, G_{yt} (Figure 5.2a), was introduced along a non-slice-selection axis (e.g., the phase-encoding axis as shown in Figure 5.2a) concurrently with the nominal slice-selection gradient (along the z-direction by convention) during the RF excitation pulse. With the G_{yt} gradient, the slice selected by the RF excitation pulse was tilted away by a small angle θ_y from the nominal slice selected by the RF refocusing pulses (Figure 5.3a,b). Since G_{yt} is concurrent with the nominal slice-selection gradient, the amplitude of the nominal gradient must be adjusted in order to maintain the prescribed slice thickness. The resultant slice-selection gradient G_z and the tilt gradient G_{yt} were calculated by solving the simultaneous equations:

$$\theta_y = \arctan(G_{yt} / G_z) \quad [5.1]$$

$$G_s^2 = G_{yt}^2 + G_z^2 \quad [5.2]$$

where G_s is the amplitude of the original slice-selection gradient. For off-centered slices, the transmitter frequency was adjusted accordingly to account for the frequency offset arising from the net slice selection gradient, G_s .

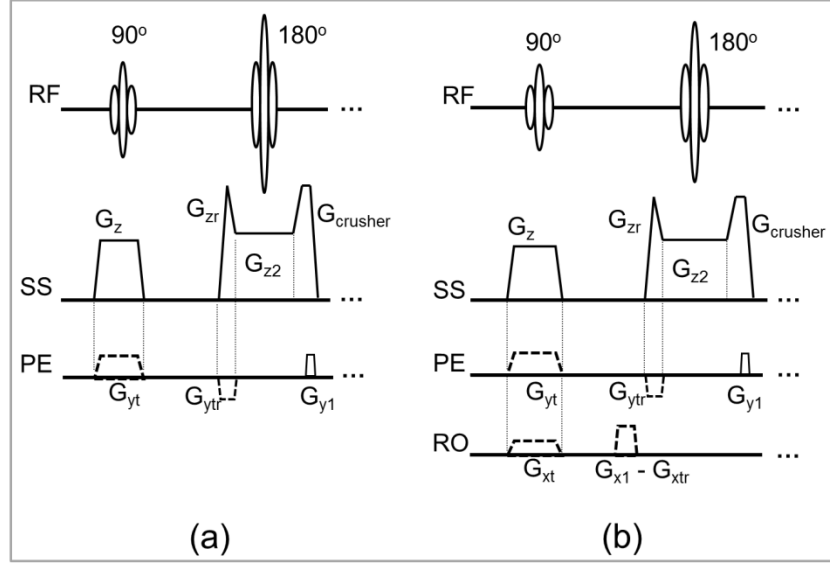


Figure 5.2: A segment of the modified FSE pulse sequence illustrating (a) the ‘orthogonal’ slice-tilting technique, and (b) the ‘oblique’ slice-tilting technique. The tilting gradients are shown by dotted lines.

Similar to the nominal slice-refocusing gradient, an additional slice-refocusing gradient, G_{ytr} , was designed and played together with the left crusher gradient of the first RF refocusing pulse, G_{zr} (Figure 5.2a). This gradient lobe was designed to coincide with the first left crusher gradient lobe just before the 180° RF refocusing pulse. The design of G_{ytr} was governed by the following rules:

1. The total area of G_{ytr} was equal to the effective area of G_{yt} , i.e., from the isodelay point to the end of the gradient lobe.
2. The total pulse width, t_{PW} , of the G_{ytr} gradient lobe was defined equal to the total pulse width of the first left crusher lobe.
3. The pulse-widths of the increasing and decreasing ramps of G_{ytr} , called t_{ramp} , were assumed to be equal. This value was calculated using the above assumptions, as the smallest real, positive solution to the quadratic equation:

$$t_{ramp}^2 SR - t_{PW} SR t_{ramp} + A = 0, \quad [5.3]$$

where t_{ramp} is the duration of the increasing and decreasing ramps of G_{ytr} , t_{PW} is the total pulse width of G_{ytr} , A is the total area of G_{ytr} , and SR is the selected slew rate of the gradient system.

4. The amplitude, a , of the gradient lobe G_{ytr} was then calculated as:

$$a = \frac{A}{t_{PW} - t_{ramp}} \quad [5.4]$$

Note that one of the design factors for G_{ytr} is the slew rate of the system, which can be user-defined as long as it is within the specified limits of the system hardware. These constraints (e.g., pulse width, slew rate, and gradient amplitude) on the design of G_{yt} and G_{ytr} imposed a limit on the maximal value of the user-selectable tilt angle θ_y , beyond which G_{ytr} was unable to completely rephase the spins. This maximum allowable tilt angle was found to be rather large on our system. For example, the maximal tilt angle was more than 70° for a slice thickness of 5 mm with the slew rate set to 90% of the maximum (150 T/m/s) of our scanner. This tilt angle limit was considerably larger than any tilt angle required for practical implementation of the technique (see Figure 5.6 in Section 5.3).

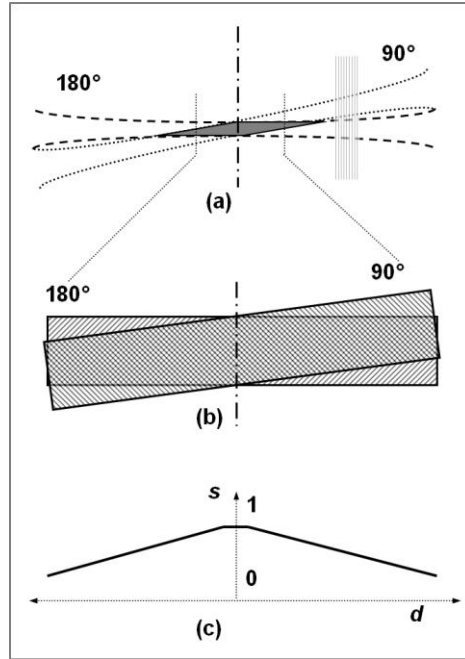


Figure 5.3: (a) Slice profiles for the 90° and 180° RF pulses in FSE, conceptually showing the overlapping region (shaded) and the location of the source of the artifact (vertical lines). The profiles of the slices reflect the non-linearity of the gradient. (b) Zoomed profile of slice overlap for the 90° and 180° pulses within the FOV. (c) Normalized theoretical signal loss s (vertical axis) along the slice tilt direction d (horizontal axis).

Although Figure 5.2a illustrates G_{yt} only along the phase-encoding direction, the tilt can also be applied along the readout direction or both the readout and phase-encoding directions, as shown in Figure 5.2b. When the slice-tilting gradient is limited to one logical axis, we call it “orthogonal tilt”. When the slice-tilting gradient is implemented on both axes, we refer to the tilt as “oblique tilt” to facilitate the following discussions. In the case of oblique tilt, two user-defined tilt angles, θ_x and θ_y , were chosen independently along the frequency and phase-encoding axes, respectively. The “slice-selection” gradient

on the readout axis, G_{xt} , was designed in a manner analogous to G_{yt} , described by Eq. 5.1, which involved solving three simultaneous equations (Eqs. 5.1, 5.5, and 5.6), shown as:

$$\theta_x = \arctan(G_{xt} / G_z) \quad [5.5]$$

$$G_s^2 = G_{xt}^2 + G_{yt}^2 + G_z^2 \quad [5.6]$$

In our sequence design, the slice-rephasing gradient for G_{xt} was combined with the readout prephasing gradient for simplicity and improved time efficiency. In this case, the transmitter frequency also took into account the frequency offset arising from G_{xt} .

These pulse sequence modifications necessitated *no changes* in image reconstruction.

5.3 Calibration Studies

The purpose of the following experiments was to characterize the appearance and strength of the FSE cusp artifact for a given set of imaging parameters. The first parameter is slice thickness, which depends upon the strength of the slice selection gradient, and is therefore expected to impact the appearance and strength of the cusp artifact. As the slice thickness is also directly correlated with the RF bandwidth (assuming that the slice selection gradient amplitude remains unchanged), the RF bandwidth may also influence the appearance and/or the strength of the cusp artifact in the same manner as the slice thickness. Imaging parameters that influence sampling along the phase-encoding direction can also influence the appearance of the cusp artifact, although they are not expected to change the strength of the artifact, or the optimal tilt angle required for its reduction. The experimental setup described below was used to characterize the cusp artifact and find the optimal tilt angle as a function of the imaging parameters.

5.3.1 Experiments

The modified FSE pulse sequence was implemented on two GE Signa HDx MRI scanners (GE Healthcare, Waukesha, WI, USA) at 3.0 T and 1.5 T, respectively. Both scanners were equipped with a CRM (Cardiac Resonator Module) gradient subsystem (maximum gradient strength = 40 mT/m, maximum slew rate = 150 T/m/s). Using this pulse sequence, a phantom study was performed to determine the optimal tilt angle under several experimental conditions on each of the two scanners, with the experimental setup similar to that shown in Figure 5.4. On the 3.0 T scanner, an 18 cm (DSV) spherical phantom containing dimethyl silicone, gadolinium and colorant was placed inside a four-element neurovascular RF coil (USA Instruments, Cleveland, OH). On the 1.5 T scanner, a similar phantom filled with water (18 cm DSV, 3.3685 g/L $\text{NiCl}_2 \cdot 6\text{H}_2\text{O}$ and 2.4 g/L NaCl) was scanned with an eight-element neurovascular RF coil (Medrad Inc., Indianola, PA). To mimic the source of the FSE cusp artifact, a smaller (11 cm DSV, 3.3685 g/L $\text{NiCl}_2 \cdot 6\text{H}_2\text{O}$ and 2.4 g/L NaCl) water phantom was placed approximately 22 cm away (i.e., along the positive z-axis) from the isocenter. The position of this phantom was adjusted such that it produced the strongest cusp artifact over a FOV of 24 cm. With this setup (Figure 5.4), the optimal tilt angle was determined as the smallest angle that minimized both the intensity and the extent of the FSE cusp artifact while leaving the signal within the FOV minimally affected.

On each scanner, a set of calibration experiments was performed to establish the relationship between the optimal tilt angle and several scan parameters. The imaging parameters for this study were: TR = 2000 ms, TE = 10 ms, ETL = 8, bandwidth = 62.5 kHz, acquisition matrix = 256×256, NEX = 2, FOV = 24 cm, slice thickness = 5 mm, and optimal tilt angle $\theta_y = 2.0^\circ$. In the first calibration, the slice thickness was increased from 2 to 10 mm in increments of 2 mm, while keeping all other parameters the same. At each slice thickness, the optimal tilt angle was determined. The relationship between the slice thickness and the optimal tilt angle was also examined at two additional FOVs: 26 cm and 28 cm. In the second calibration, the strength of the cusp artifact was calculated with the slice thickness held constant at

5 mm, and the FOV increasing from 20 cm to 30 cm in steps of 2 cm. Third, the artifact was evaluated against readout bandwidths of 12.5 kHz, 15.63 kHz, 20.33 kHz, 31.25 kHz, and 62.5 kHz. In the fourth experiment, the influence of echo train length (ETL) on the cusp artifact and, consequently, on the optimal tilt angle, was investigated by varying the ETL from 4 to 20 in increments of 4. Finally, the artifact was evaluated against phase encoding steps from 128 to 256, in steps of 32.

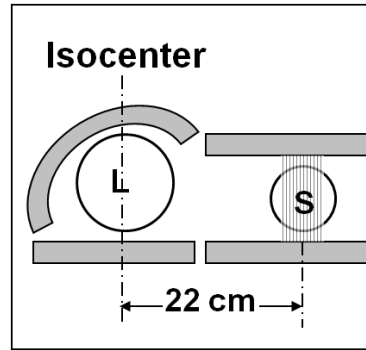


Figure 5.4: Setup of the phantom experiment using a four-channel neurovascular coil.

Once the optimal tilt angle along the phase-encoding direction, θ_y , was established for orthogonal slice tilting, the tilt angle, θ_x , was applied along the frequency-encoding direction to find the effectiveness of using oblique slice tilting.

5.3.2 Quantification

The FSE cusp artifact was first visually compared before and after applying the slice-tilting technique, followed by a quantitative analysis of the artifact strength. In the quantitative analysis, three regions of interest (ROI), each comprising ~ 100 pixels, were selected from (i) a homogeneous region

(10x10 pixels) of the object with the highest signal intensity, (ii) the background (10x10 pixels) free of artifacts, and (iii) the area with the strongest artifact strength (10x10 pixels). The strength of the artifact (s_a) was quantified using the following equation:

$$s_a = \frac{\mu_o - \mu_n}{\mu_o - \mu_n} \quad [5.7]$$

where μ_o , μ_a and μ_n were the mean signal intensity of the ROI over the object, the artifact, and the background, respectively.

5.3.3 Results

The following results show a reduction in the FSE cusp artifact when a small tilt angle is applied to the selected slice. For example, using a slice tilt of 2° for a slice thickness of 5 mm over a FOV of 24 cm, the strength of the cusp artifact was reduced from 16.5 % (Figure 5.5c) before to 3.1 % (Figure 5.5c') after applying the tilt. (Note that the typical FSE ghosting artifacts can also be seen on the lower right side of the images. These artifacts are unrelated to the FSE cusp artifact described in this paper.) Further, the amplitude of the slice selection gradient (G_s), which is inversely proportional to the slice thickness, also determines the strength of the cusp artifact, and influences the optimal tilt angle for that slice thickness. Figure 5.5a–f shows the decrease in the artifact strength from 26.3 % to 9.9 % as the slice thickness increases from 2 mm to 10 mm.

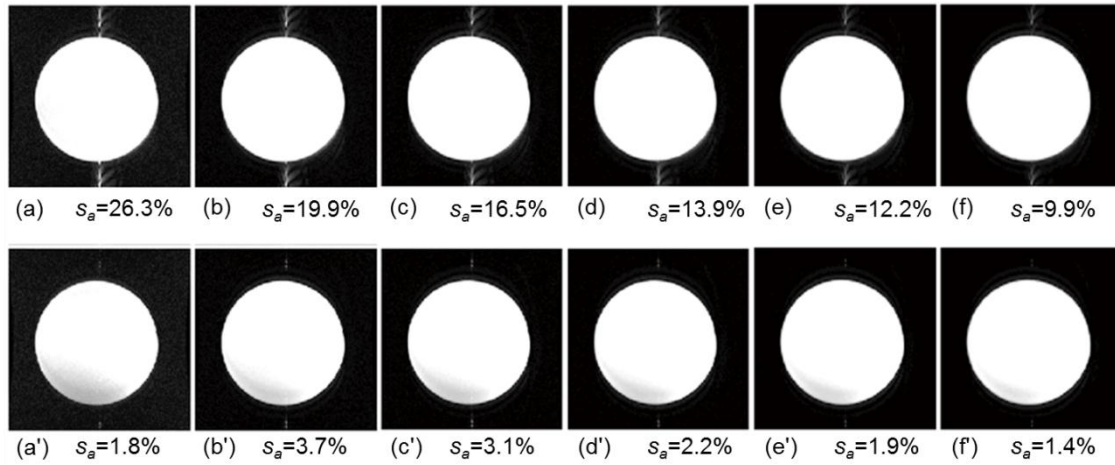


Figure 5.5: The cusp artifact is shown with a varying slice thickness of (a) 2 mm, (b) 4 mm, (c) 5 mm, (d) 6 mm, (e) 8 mm, and (f) 10 mm. (a' – f') are the corresponding images when the slice is tilted by its corresponding optimal tilt angle (Figure 5.6).

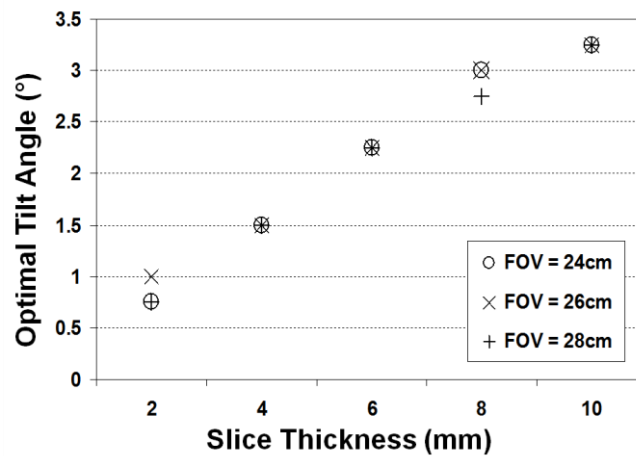


Figure 5.6: Optimal tilt angle θ_y as a function of slice thickness for three different FOVs.

Figure 5.6 shows the optimal tilt angle obtained from the phantoms on the 3.0 T scanner as a function of the slice thickness for a FOV of 24 cm (shown by the ‘o’ signs). The corresponding images using the optimal tilt angles for the respective slices are shown in Figure 5.5a’–f’. The artifact strength decreases to an average of $s_a = 2.3\%$ after the slice tilting was applied. The optimal tilt angle was also calculated for FOVs of 26 cm and 28 cm, shown respectively by the ‘x’ and ‘+’ signs in Figure 5.6. In all cases, the relationship between the slice thickness and the optimal tilt angle was found to be approximately linear (e.g., $r = 0.9883$ for FOV = 24 cm). The same relationship was confirmed for human volunteers, suggesting that the optimal tilt angles obtained on a phantom scan were usable for other scans with the same or similar slice thickness.

The cusp artifact was not shown to have a significant dependence on the FOV, as illustrated by Figure 5.7. The average strength of the cusp artifact was $s_a = 22.4\%$. The dependence of the optimal tilt angle on the FOV was also very weak, as illustrated by Figure 5.6.

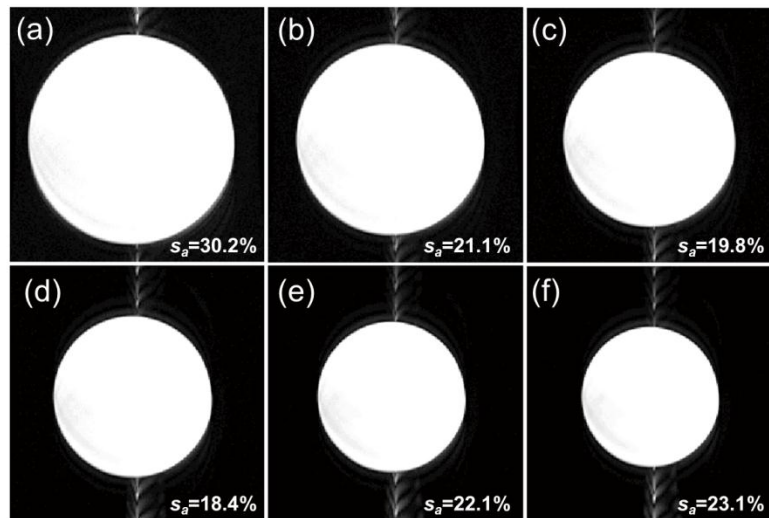


Figure 5.7: The strength of the FSE cusp artifact was not shown to be dependent upon FOVs of: (a) 20 cm, (b) 22 cm, (c) 24 cm, (d) 26 cm, (e) 28 cm, and (f) 30 cm, acquired with slice thickness = 5 mm.

Although the readout bandwidth was not expected to influence the optimal tilt angle, its impact on the strength and appearance of the artifact was studied in the third experiment. Figure 5.8 shows that the artifact strength is independent of the readout bandwidth and, consequently, the spacing between echoes. Similar results were obtained for readout bandwidths of 12.5 kHz and 20.83 kHz (images not shown). The optimal tilt angle was independent of the readout bandwidth.

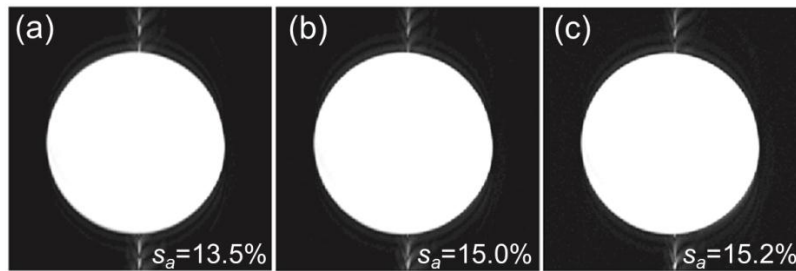


Figure 5.8: The cusp artifact is independent of the readout bandwidths, for the images shown with bandwidths of (a) 15.63 kHz, (b) 31.25 kHz, and (c) 62.5 kHz. These images were acquired with slice thickness of 5 mm and FOV of 24 cm.

The final two calibration experiments using orthogonal slice-tilting were performed to determine the dependence of the artifact, and the optimal tilt angle, on the ETL and number of phase-encoding steps. Figure 5.9a–c show that the strength of the artifact increases as the ETL increases. The strength and width of the artifact was also shown to decrease as the total number of phase-encoding steps increases, in Figure 5.10a–c. In both cases, the optimal tilt angle was found to be independent of either ETL (Figure 5.9a'–c') or phase-encoding steps (Figure 5.10a'–c').

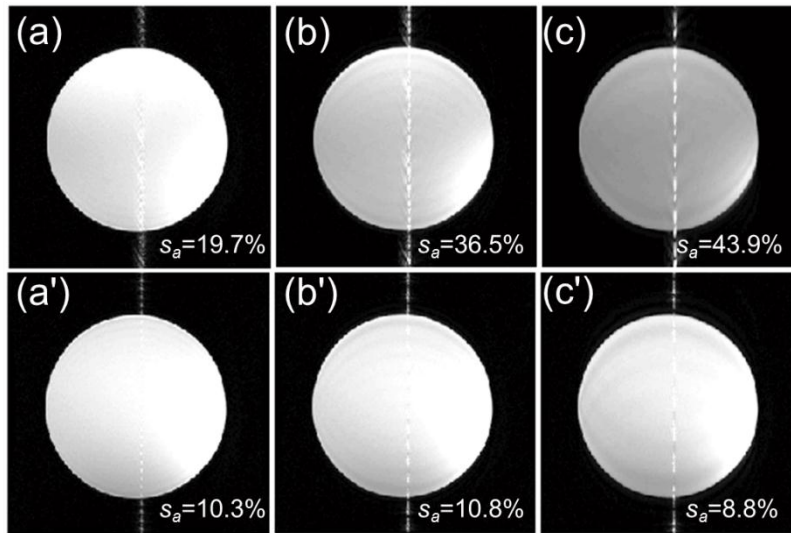


Figure 5.9: The strength of the artifact increases as ETL increases from (a) 4, and (b) 8, to (c) 16. The corresponding images obtained with the optimal tilt angle of 2° (for all studies) are shown in (a')–(c').

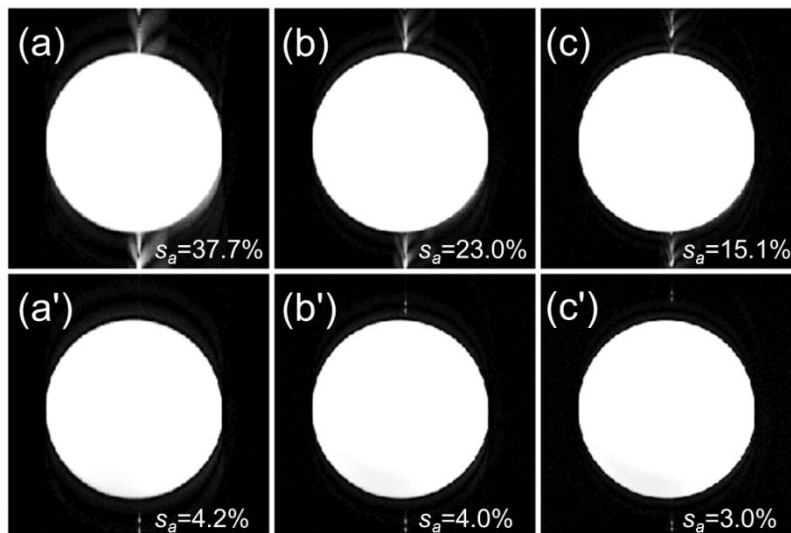


Figure 5.10: The strength of the artifact decreases as the number of phase-encoding steps increases, from (a) 128, and (b) 192, to (c) 256. The corresponding images obtained with the optimal tilt angle of 2° (for all studies) are shown in (a')–(c').

Together, the choice of ETL and phase-encoding steps determine the number of excitations required for image acquisition, i.e., the total number of shots is higher when smaller ETL or more phase-encoding steps are selected. After the slice-tilting is applied, the cusp artifact takes on a very well-defined appearance on the phantom image. It can be observed from Figure 5.9 and Figure 5.10 that the number of excitations influences the appearance of the cusp artifact.

All of the results discussed above were also obtained on the 1.5 T scanner (data not shown) with the same gradient sub-system.

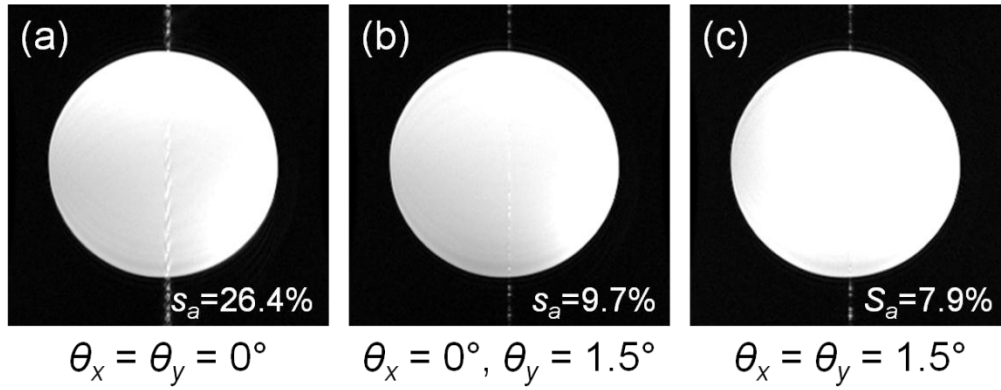


Figure 5.11: The orthogonal slice-tilting technique is compared with the oblique slice-tilting technique. The phantom image is shown with (a) $\theta_x = \theta_y = 0^\circ$, (b) the orthogonal tilt, with $\theta_x = 0^\circ, \theta_y = 1.5^\circ$, and (c) the oblique tilt, with $\theta_x = \theta_y = 1.5^\circ$.

Although the orthogonal slice-tilting approach demonstrated considerable reduction of the cusp artifact, a residual artifact remains after applying the tilt to the phantom images. For this reason, the oblique tilt was applied to assess its efficiency in further reducing the artifact. For the imaging parameters used in this study, the optimal tilt angle along the readout axis was also calculated as $\theta_x = 1.5^\circ$. Figure

5.11 shows a comparison of the orthogonal slice-tilting approach with the oblique slice-tilting approach. The artifact decreases from 26.4 % before to 9.7 % after applying a tilt of $\theta_y = 1.5^\circ$, and 7.9 % after applying a tilt of $\theta_x = \theta_y = 1.5^\circ$. A comparison between Figure 5.11b and Figure 5.11c also indicates no significant visual improvement by using the oblique tilt over the orthogonal tilt. For this reason, all the experiments in the following section used the orthogonal slice-tilting approach.

5.4 Validation Experiments

After the optimal tilt angle was determined under different experimental conditions, four experiments were conducted on phantoms and human volunteers to demonstrate and evaluate the performance of the proposed technique. The first experiment was carried out on phantoms at 3.0 T with the same setup as described in Figure 5.4, except that the 18 cm DSV spherical phantom at the isocenter was replaced with the structured phantom recommended by the ACR (American College of Radiology). The ACR phantom was scanned with a sagittal cervical spine protocol that we used clinically. The key acquisition parameters were: TR = 2000 ms, TE = 10 ms, ETL = 8, bandwidth = 62.5 kHz, acquisition matrix = 256×256, NEX = 2, FOV = 24 cm, slice thickness = 5 mm, and optimal tilt angle $\theta_y = 2.0^\circ$.

The second experiment was to validate the phantom results on human subjects at 3.0 T. The same neurovascular RF coil used in the phantom experiment was employed to obtain images from the left foot of two healthy human subjects (a 26-year old male and a 31-year old female) under an approved IRB protocol. By using the same coil, results from the *in vivo* human studies can be compared directly with those from the phantom experiments. With the subject in a supine position, a T_1 -weighted sagittal foot examination was performed with the following imaging parameters: TR = 600 ms, TE = 20 ms, ETL = 8, bandwidth = 62.5 kHz, acquisition matrix = 256×256, NEX = 4, FOV = 26 cm, slice thickness = 5 mm, and optimal tilt angle $\theta_y = 2.0^\circ$. The optimal tilt angle, obtained from the phantom calibration described earlier, was confirmed in one of the human volunteer studies. The artifact-producing region for the *in vivo* studies was found to be 22.5 ± 1.0 cm away from the isocenter, i.e., the same as in the phantom

experiment illustrated in Figure 5.4. The results from the two volunteers were compared to evaluate the performance consistency of the slice-tilting technique.

In the third experiment, we evaluated the performance of the slice-tilting technique in one of our clinical protocols in which the FSE cusp artifact was most problematic. Sagittal images of the thoracic spine were acquired from a 26-year old male volunteer on the 1.5 T GE Signa scanner using an eight-element cervical-thoracic-lumbar (CTL) spine coil provided by the equipment manufacturer. The imaging parameters included: TR = 3500 ms, TE = 120 ms, ETL = 16, bandwidth = ± 15.63 kHz, acquisition matrix = 256x256, NEX = 2, FOV = 24 cm, slice thickness = 5 mm, and optimal tilt angle $\theta_y = 2.0^\circ$ with active elements 3 through 5 (Figure 5.12). Since tilting the slice can potentially affect neighboring slices, three contiguous slices without any inter-slice gap were acquired to evaluate the performance of the slice-tilting technique for multi-slice imaging.



Figure 5.12: Setup of the sagittal thoracic spine experiment showing the approximate locations of the active elements (3–5) of the CTL coil.

The fourth experiment was designed to compare the slice-tilting technique with two commercial techniques available on our 3.0 T scanner for FSE cusp artifact reduction. Both commercial techniques were based on a common principle of shifting the source of the cusp artifact. The first technique swapped the position of the artifact source during slice selection, by reversing the direction of the slice-selection gradient applied during the 90° RF excitation pulse relative to the slice-selection gradient for the refocusing pulses. The second technique employed a different bandwidth between the RF excitation and refocusing pulses, consequently using a different slice-selection gradient amplitude to dislocate the source of the artifact. In our experimental studies, the bandwidth ratio between the RF excitation and refocusing pulses was varied from 0.5 to 2.0 in steps of 0.5, while keeping the slice thickness constant at 5 mm. Both techniques were evaluated on the left foot of a human volunteer with the same set-up and protocol as described in the second experiment.

The strength of the FSE cusp artifact was measured as discussed in Section 5.35.3.2. In the quantitative analysis, three regions of interest (ROI), each comprising ~ 100 pixels, were selected from (i) a homogeneous region (10×10 pixels) of the object with the highest signal intensity (i.e., on the calcaneus of the foot of the volunteer, or on the subcutaneous fat of the thoracic images from the volunteer), (ii) the background (10×10 pixels) free of artifacts, and (iii) the area with the strongest artifact strength (10×10 or 20×5 pixels, depending upon the appearance of the artifact). The strength of the artifact (s_a) was quantified using Eq. 5.7.

5.5 Results

A representative result from the first (phantom) experiment is shown in Figure 5.13. The cusp artifact is enclosed in a dashed box on the images. By applying a tilt of 2° for a slice thickness of 5 mm over a FOV of 24 cm, the strength of the cusp artifact was reduced from 9.0% (Figure 5.13a) to 4.6% (Figure 5.13b). As a result of this substantial reduction, the diffuse artifact structure virtually disappeared and the width of the remaining artifact was noticeably narrowed.

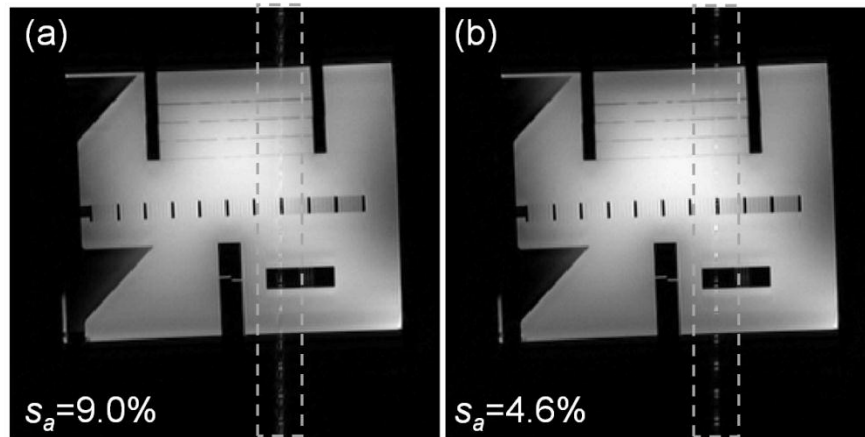


Figure 5.13: FSE images of the ACR phantom (a) without slice tilting, and (b) after applying a tilt angle of 2° .

The results from the second experiment on human foot images at 3.0 T are shown in Figure 5.14 where an unconventional window/level setting was used to emphasize the artifacts. On one of the volunteers, the artifact strength was reduced from 3.0% (Figure 5.14a) before to 0.5% (Figure 5.14b) after applying a tilting angle of 2.0° . The artifacts in the resultant image were essentially invisible even with a window and level setting to highlight the low signal intensities. Results from the second human volunteer showed a similar performance, with the artifact strength decreasing from 5.1% to 0.9% before and after the tilt (images not shown), demonstrating good inter-subject consistency of the slice-tilting technique.

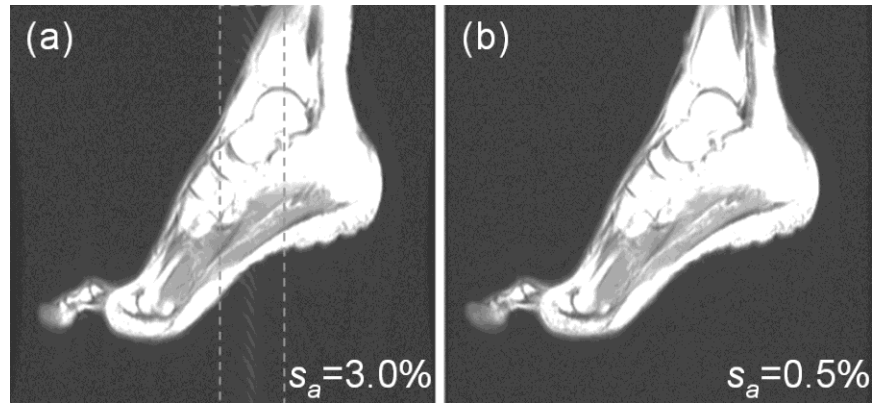


Figure 5.14: FSE images of human foot at 3.0 T (a) without slice-tilting, and (b) after slice tilting by 2° .

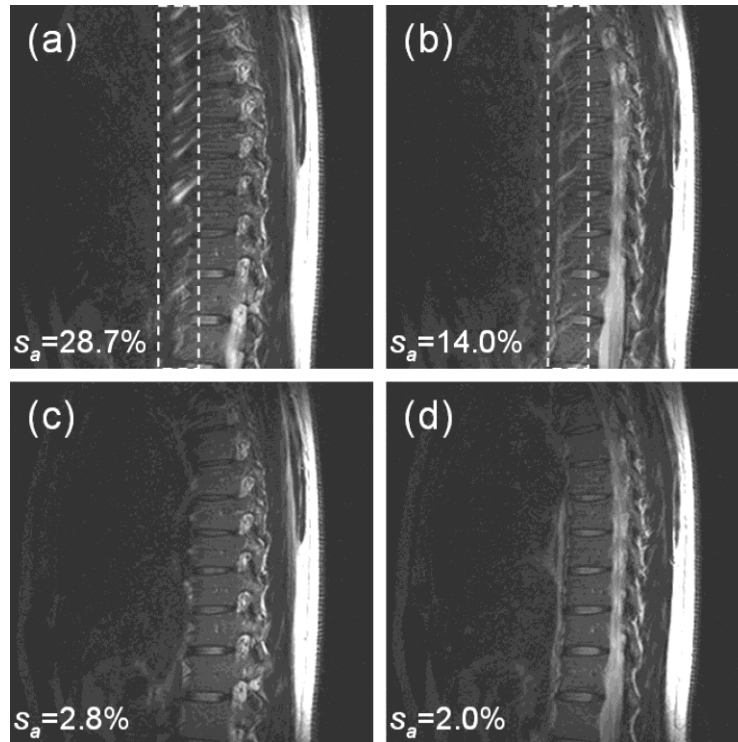


Figure 5.15: Two adjacent sagittal images of thoracic spine at 1.5 T (slice thickness = 5 mm; slice gap = 0 mm) (a-b) without slice-tilting and (c-d) with slice tilting ($\theta_y = 2^\circ$).

An even better performance was obtained in the human thoracic spine study (i.e., the third experiment) at 1.5 T. Figure 5.15 shows two of the three contiguous sagittal slices before (Figure 5.15a,b) and after (Figure 5.15c,d) applying the slice-tilting technique. Before tilting, the characteristic feather-like artifacts were evident along the superior-inferior direction (i.e., the phase-encoding direction) with artifact strength as high as 14.0% (Figure 5.15b) to 28.7% (Figure 5.15a). After tilting, the artifact strength was reduced to 2~3% (Figure 5.15c,d), which was virtually invisible even with a window and level setting to emphasize the artifact. It is worth noting that no significant image shading was observed across the FOV after applying the slice-tilting technique. Additionally, the effect of cross-talk in the multi-slice acquisition was insignificant even with a tilt angle of 2° without any inter-slice gap.

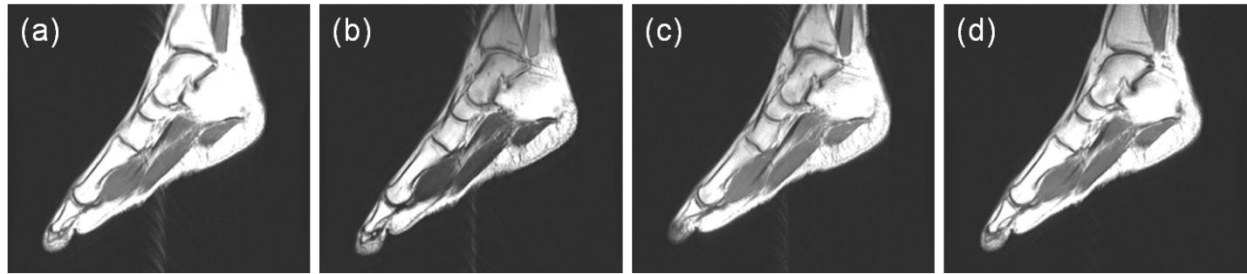


Figure 5.16: Sagittal foot FSE images from a human volunteer comparing the slice-tilting technique with two commercially available techniques. (a) Original image; (b) Image obtained by reversing the direction of the slice-selection gradient during the RF excitation pulse; (c) Image obtained by increasing the bandwidth of the RF excitation pulse to twice of that of the RF refocusing pulse; (d) Image with slice-tilting ($\theta_y = 2^\circ$).

Figure 5.16 summarizes the results of the fourth experiment which compares the slice-tilting technique with two commercially available techniques. Figure 5.16a displays the original image without

any compensatory techniques. Figure 5.16b and Figure 5.16c show the result of reversing the slice-selection gradient, and varying the bandwidth of the RF excitation pulse to twice the original bandwidth (a bandwidth ratio of 2 was experimentally observed to give the best artifact reduction for this technique), respectively. The result of the slice-tilting technique (tilt angle = 2°) is given in Figure 5.16d. Visual inspection of these images clearly showed that the slice-tilting technique outperformed the two commercial techniques.

5.6 Optional Post-Processing

This section discusses the use of data-processing approaches after the slice tilting technique reduced the strength and width of the cusp artifact. Two post-processing approaches were introduced for: (a) the reduction of the ‘residual’ artifact, especially seen on the phantom images, and (b) improving the uniformity of the image, since the tilt angle can reduce the signal intensity at the edge of the FOV.

5.6.1 *Reduction of Residual Artifact*

By applying a tilt of 2° for a slice thickness of 5 mm over a FOV of 24 cm, the strength of the cusp artifact was reduced from 16.3% (Figure 5.17a) to 5.3% (Figure 5.17b) in the phantom images. However, in most phantom experiments, it was observed that a minor residual artifact (i.e., a thin line in the phase encoding direction with a width no more than 2-4 pixels) remained after the slice-tilting technique substantially reduced both the intensity and the width of the cusp artifact (Figure 5.17b). To further reduce this residual artifact, we implemented a simple interpolation algorithm which involves the following steps. First, a line containing the residual artifact (i.e., two to four pixels) was zeroed out along the phase-encoding direction of the image. For example, the following operation was executed assuming that the artifact line was three-pixels wide and at the center of the FOV:

$$I(i, j) = 0, \quad 1 \leq i \leq R, \quad \frac{C}{2} - 1 \leq j \leq \frac{C}{2} + 1 \quad [5.8]$$

where $I(i, j)$ denotes the image intensity of the i th row (i.e., readout direction) and the j th column (i.e., the phase-encoding direction), R is the total number of rows, and C is the total number of columns of the image. Then, along each row, the missing signal intensities were linearly interpolated from the two immediately adjacent pixels. In the example given in Eq.5.8, the interpolation was performed in the following manner:

$$\begin{bmatrix} I(i, \frac{C}{2} - 1) \\ I(i, \frac{C}{2}) \\ I(i, \frac{C}{2} + 1) \end{bmatrix} = \frac{1}{4} \begin{bmatrix} 1 & 3 \\ 2 & 2 \\ 3 & 1 \end{bmatrix} \begin{bmatrix} I(i, \frac{C}{2} + 2) \\ I(i, \frac{C}{2} - 2) \end{bmatrix}, \quad 1 \leq i \leq R \quad [5.9]$$

The use of this interpolation algorithm is optional, depending on the presence and severity of the residual artifact. In the experiments described in Section 5.5, only the first experiment using phantoms showed a residual artifact after application of the slice tilting. Figure 5.17 shows the result of applying the post-processing step to reduce the artifact further. The artifact strength decreased further from 5.3 % after using a tilt angle of 2° (Figure 5.17b) to 1.8 % after the additional interpolation step was applied (Figure 5.17c).

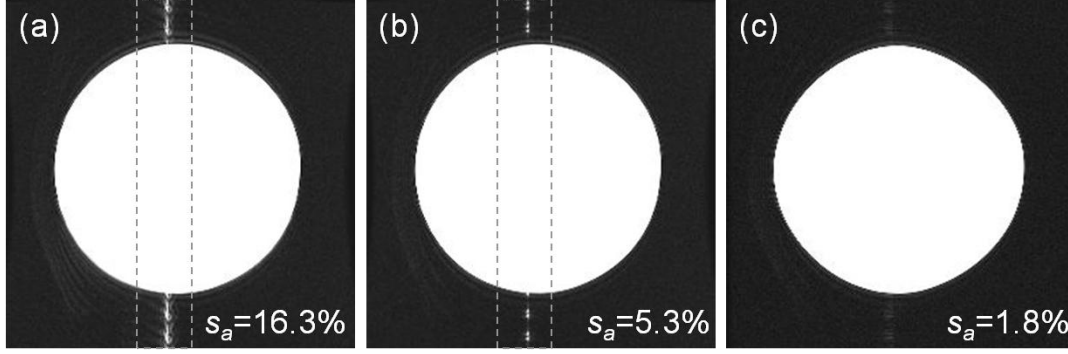


Figure 5.17: FSE images of an 18cm (DSV) phantom (a) with tilt angle = 0° , (b) with tilt angle = 2° , and (b) with tilt angle = 2° after post-processing described by Eqs. 5.8 and 5.9.

5.6.2 The Signal Intensity Change

The incomplete overlap between the slices selected during the excitation and refocusing pulses will cause the signal intensity at the edge of the FOV to be lower than that at the center. The loss in signal intensity (s , see Figure 5.3c) due to slice tilting can be theoretically described by the following equation:

$$s = \begin{cases} 1 & 0 < d \leq \tan(\theta/2)th/2 \\ -d \cdot \tan(\theta/2) + (1 + \sec \theta)/2 & \tan(\theta/2)th/2 < d \leq \cot(\theta/2)th/2 \\ 0 & \cot(\theta/2)th/2 < d \end{cases} \quad [5.10]$$

where th is slice thickness, d is the distance from the center of the FOV, and θ is the applied tilt angle. Using this equation, the theoretically calculated signal intensity at the edge of the FOV was only a small fraction (e.g., $\sim 20\%$) of the signal without the tilt, for a FOV of 24 cm, with a 5 mm slice thickness and a tilt angle of 2° . However, this calculation assumed (i) an ideal gradient linearity, (ii) a uniform RF coil sensitivity profile across the FOV, and (iii) a homogeneous signal intensity distribution within the slice. To quantify the actual signal intensity drop at the edge of the FOV as a result of slice tilting, an experiment was conducted on the 3.0 T scanner using the body coil for both RF transmission and

reception. Images were acquired from a large (~26 cm DSV) spherical phantom containing dimethyl silicone with the following parameters: TR = 2000 ms, TE = 10 ms, ETL = 16, bandwidth = 62.5 kHz, acquisition matrix = 256×256, NEX = 2, FOV = 26 cm. The slice thickness was held constant at 2 mm while the tilt angle was increased from 0° to 3° in steps of 1°. This experiment was repeated with slice thickness of 4 mm, 5 mm, 6 mm, 8 mm and 10 mm, for single-slice and multi-slice (comprising three slices without inter-slice gap) acquisitions, respectively.

Signal uniformity was evaluated in the direction of the slice tilt (i.e., the phase-encoding direction) for all images acquired in the fifth experiment. Two ROIs, each comprising ~100 pixels, were selected from (i) the highest signal region (10x10 pixels) at or near the center of the spherical phantom, and (ii) the lowest signal region (10x10 pixels) within 2.5 cm of the edge of the FOV on the phantom in the direction of the tilt. The signal uniformity, U , was calculated using the following equation:

$$U = 1 - \frac{\mu_c - \mu_e}{\mu_c + \mu_e} \quad [5.11]$$

where μ_c and μ_e were the mean signal intensity of the ROIs taken at the center and the edge, respectively.

The results of the signal uniformity evaluation on the phantom are summarized in Figure 5.18. Figure 5.18a shows the values of percent signal uniformity as a function of slice thickness and tilt angle for single-slice imaging. For example, for a slice thickness of 5 mm and a tilt angle of 2°, the intensity uniformity was found to be 71%. The uniformity values for single-slice imaging are comparable to those in multi-slice imaging (Figure 5.18b), suggesting that the slice-tilting technique has minimal adverse effect in multi-slice acquisitions. Figure 5.18c compares the profiles of signal intensity for tilt angles from 0° to 3° for a 5 mm slice (single-slice acquisition) along the central column in the phase encoding direction. For a typical tilt angle (2°) used in all experiments, the worst signal drop was about ~40% as compared to the signal at the center. The dependence of the signal loss on slice thickness is shown in Figure 5.18d where the tilt angle was held constant at 2° in a single-slice acquisition. The signal loss became progressively worse as the slice thickness decreased, especially below 4 mm. However, as the

slice becomes thinner, the required tilt angle also decreases (Figure 5.6), which substantially compensates for the signal loss illustrated in Figure 5.18a,b,c.

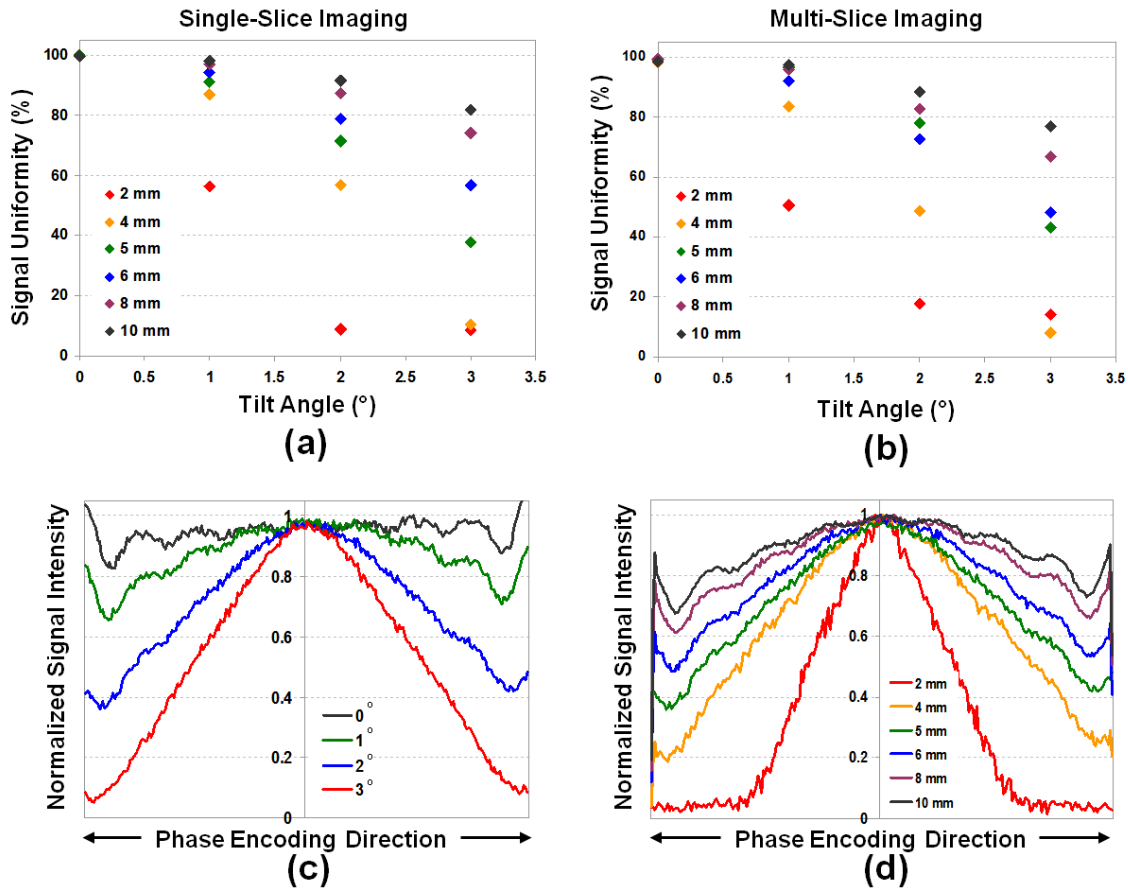


Figure 5.18: Signal non-uniformity in the direction of the slice tilt (i.e., the phase-encoding direction). (a) Signal uniformity as a function of slice thickness in single-slice imaging ($\theta_y = 0-3^\circ$); (b) Signal uniformity as a function of slice thickness in multi-slice imaging (central slice, slice gap = 0 mm); (c) Normalized signal intensity profiles in the phase-encoding direction as a function of tilt angle (slice thickness = 5 mm; single-slice acquisition); (d) Normalized profiles of signal intensity in the phase-encoding direction as a function of slice thickness (tilt angle = 2° ; single-slice acquisition).

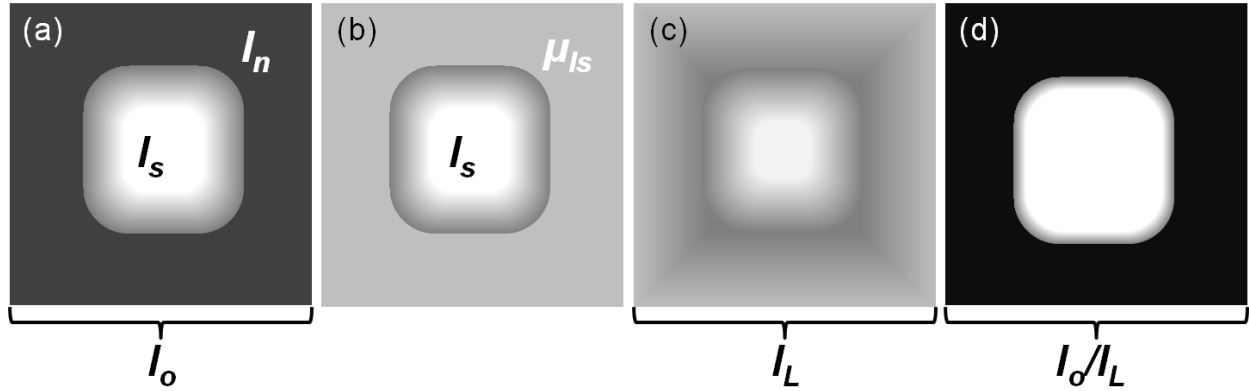


Figure 5.19: The surface coil intensity correction method first presented in (131).

Several techniques may be used to compensate for the signal intensity change. Primary among them is the class of surface-coil intensity correction methods (24,132), which have been used extensively to correct for signal non-uniformities associated with B_1 field inhomogeneity. One of the correction techniques (131) was adapted to the FSE images acquired with image non-uniformity. The algorithm was as follows:

1. Threshold the original image (I_o) to separate the object signal (I_s) from the background (I_n) (Figure 5.19a).
2. Set each pixel of I_n to the average of all pixel intensities of I_s (Figure 5.19b).
3. Smooth the entire image with the chosen kernel, call the low-pass filtered image I_L (Figure 5.19c).
4. The corrected image is obtained as a pixel-by-pixel ratio of the original image I_o to the smoothed image I_L (Figure 5.19d).

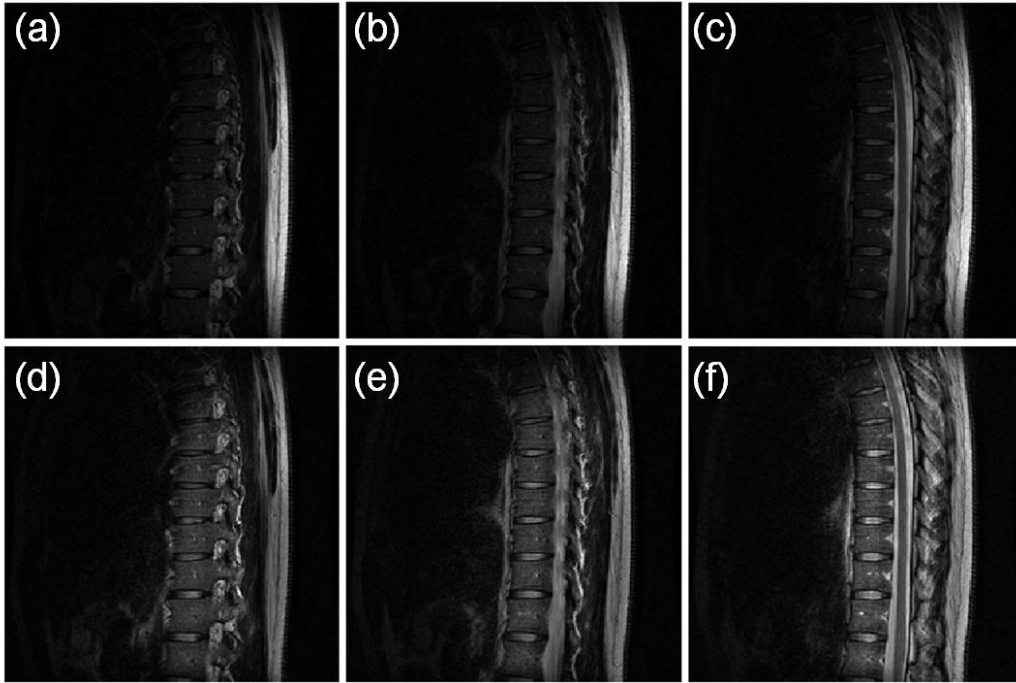


Figure 5.20: Multi-slice images (see Experiment 3) acquired at 1.5 T using $\theta_y = 2^\circ$, (a – c) without intensity correction, and, (d – f) with the SCIC technique described above. The window and level settings are the same for both sets of images.

The application of this technique to the spine images acquired on the 1.5 T scanner is shown in Figure 5.20. Figure 5.20a-c are the three contiguous slices acquired with a tilt angle of 2, of which Figure 5.20a and b are the same as Figure 5.15c and d, respectively. The images were corrected using a threshold of 5% of the highest image intensity, and a low-pass filtering kernel with a linear dimension of 25 pixels. The signal uniformity shows a marked visual improvement, seen in Figure 5.20d-f.

Although this SCIC technique provides a simple way to reduce signal non-uniformity, it suffers from a disadvantage in that the noise increases along with the signal. This means that both parameters, the kernel and the threshold, must be selected carefully in order to improve the signal uniformity while maintaining adequate SNR on the shaded regions. This may result in the loss of some information in

regions where the signal is very much lower than average. Several other image-processing techniques may be used (24) to perform the intensity correction in a more efficient manner. These techniques may also benefit from taking into account the profile of signal intensity across the FOV (Eq. 5.10). In any event, SCIC techniques are usually applied only when the shading effect is far more predominant than in the cases shown here.

5.7 Other Applications: Aliasing Artifact in FSE-PROPELLER

The slice-tilting technique is similar to several inner-volume imaging approaches (34,130), which were developed to reduce aliasing artifacts when the FOV is smaller than the object; as such, this technique can also be used to reduce aliasing artifacts in MRI, e.g., in PROPELLER sampling. In PROPELLER, the phase-encoding axis is not fixed to a specific direction. When the phase-encoding direction is rotated to the superior/inferior direction in sagittal or coronal planes, aliasing can occur in some blades, producing non-localized streaking artifacts across the image. In this study, the slice-tilting technique was applied towards the reduction of the streaking artifacts in PROPELLER imaging. As diffusion-weighted imaging in non-axial planes is preferred in some applications (e.g., visualization of corticospinal tracts using DTI), the feasibility of using this technique to obtain high-resolution DW images in sagittal and coronal planes with improved image quality has been demonstrated.

5.7.1 Methods

Similar to the theory described in 5.2, a DW FSE-PROPELLER sequence was modified to introduce a small slice-selection gradient (G_{yt}) in the phase-encoding direction (in this case, the short axis of the blade) during the 90° excitation RF pulse (Figure 5.21). With this gradient, the slice selected by the 90° pulse is tilted by an angle given by Eq. 5.1 from the prescribed slice orientation. Due to the tilt, spins excited during the 90° pulse do not overlap completely with those selected by the RF refocusing

pulses (Figure 5.3b). With an optimally selected tilt angle for a given slice thickness, signal contributions from the regions beyond the FOV can be reduced while the signals within the prescribed FOV are minimally affected. For a given FOV and slice thickness, the tilt angle was chosen such that there was minimal signal contribution from aliasing-prone regions beyond the FOV. The optimal tilt angle under most imaging conditions was rather small (e.g., $\sim 2^\circ$ for a slice thickness of 5mm).

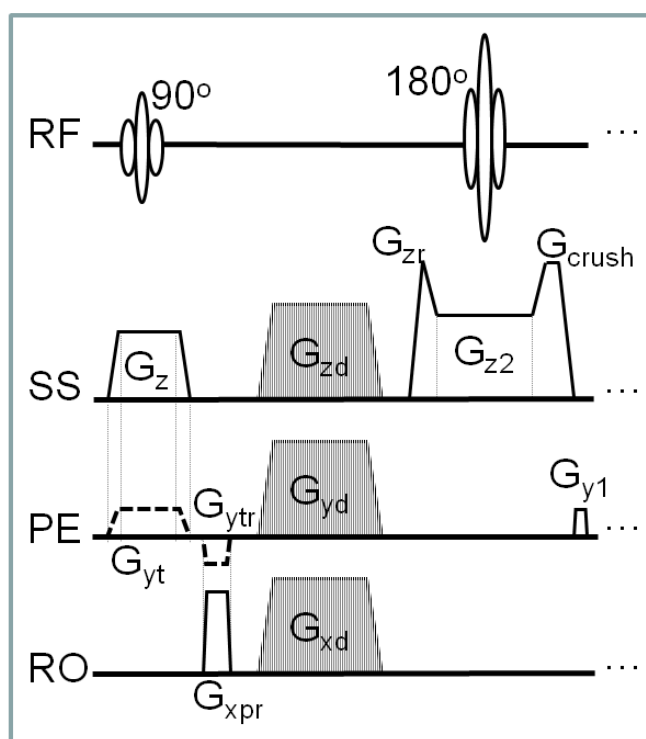


Figure 5.21: The slice tilting technique applied to reduce streaking artifacts in FSE-PROPELLER.

Experimental studies were carried out on a 1.5 T GE Signa HD scanner (GE Healthcare, Waukesha, WI). Sagittal and coronal DW images were acquired on both phantoms and a human volunteer to demonstrate the performance of the technique. The acquisition parameters included: TR/TE

= 4000/93ms, ETL = 16, bandwidth = ± 125 kHz, 48 blades, 256 readout points, FOV = 28cm, slice thickness = 5mm, and tilt = 2° . Diffusion weighting was applied along the anterior/posterior direction for the sagittal images, and along the right/left direction for the coronal images, both with $b = 750$ s/mm².

The loss in signal intensity (s ; shown in Figure 5.3c) due to slice tilting was described by Eq. 5.10. To compensate for the signal loss, a simple correction kernel (k) was derived from Eq. 5.10, and given by:

$$k = 1 - s \quad [5.12]$$

This kernel was applied on a point-by-point basis to each blade of the dataset after motion and phase correction (30), but before the final gridding reconstruction. To quantitatively evaluate artifact reduction, the strength of the streaking artifacts was calculated as a ratio of the mean intensity of a region of interest (ROI, ~ 16 pixels) in the background to an ROI in the corpus callosum before and after slice tilting and intensity compensation.

5.7.2 Results

Figure 5.22 illustrates the performance of the proposed technique for sagittal (top row) and coronal (bottom row) DW images. Figure 5.22a–d are displayed with window and level settings that highlight the streaking artifacts. The strength of the artifacts decreased from 7.3% (Figure 5.22a) and 9.2% (Figure 5.22b) before the tilt, to 2.5% (Figure 5.22c) and 3.8% (Figure 5.22d) after applying the tilt of 2° and the signal intensity correction. The images after artifact reduction (Figure 5.22c–d) are also displayed with standard window and level settings in Figure 5.22e–f.

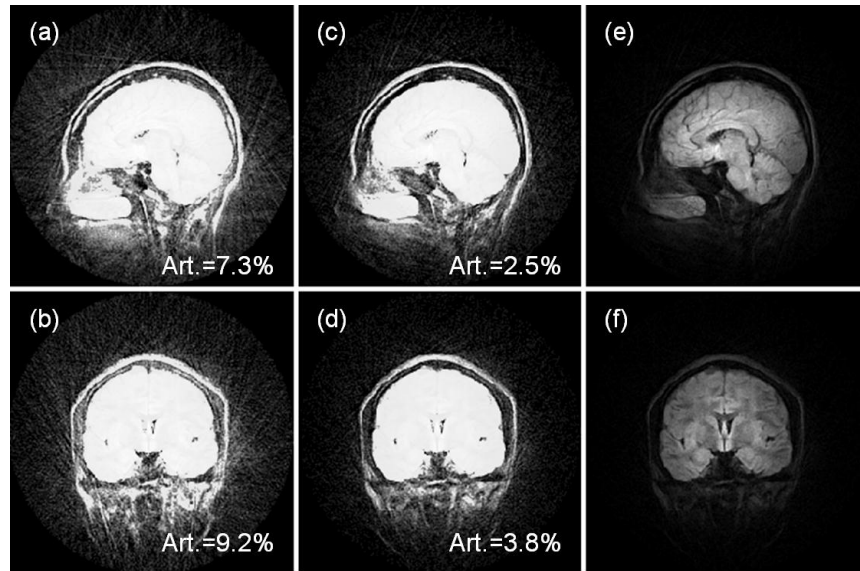


Figure 5.22: Sagittal (Top) and Coronal (bottom) DW images: (a-b) without tilt, with window and level adjusted to view artifacts; (c-d) Results of applying a tilt of 2° and the post-processing step; (e-f) The images from (c) and (d) at standard window and level settings.

5.8 Discussion

In this study, we have observed that a small slice-tilting gradient along a non-slice-selection axis during RF excitation can significantly reduce the intensity (by $\geq 50\%$) and the extent of the FSE cusp artifact. The optimal tilt angle can be determined by using a simple calibration procedure on a phantom and applied to a broad range of protocols encountered in clinical imaging. The dependence of the optimal tilt angle on slice thickness, FOV, and ETL has been studied, which revealed a linear relationship with respect to slice thickness and no significant dependence on FOV or ETL. This technique was validated on scanners operating at two different field strengths, multiple RF coils, a number of clinical protocols, and both phantoms and human volunteers. Consistent and reproducible artifact reduction was observed in all cases.

Compared to existing methods for FSE cusp artifact reduction, the slice-tilting technique offers a number of advantages. First, unlike the APA method, the slice-tilting technique requires no change in the system hardware. It can be implemented on virtually all scanners where the FSE cusp artifact is problematic. Second, the slice-tilting technique imposes no changes in image reconstruction algorithms. This is in sharp contrast to the parallel imaging approaches in which the coil sensitivity profiles must be incorporated into image reconstruction (126,128). Third, the calibration procedure in the slice-tilting technique is simple and straightforward. It does not require an additional RF coil to acquire signals from the artifact-prone region and one calibration can be used for a number of clinical protocols. Since the optimal tilt angle spans a very small range under most circumstances, the calibration procedure may even be eliminated by using a generic tilt angle in the range of 1-3°, as further discussed below. Fourth, compared with methods such as the application of an RF blanket, the slice-tilting technique imposes neither additional safety concerns nor changes in patient handling. Lastly, the slice-tilting technique has exhibited better performance in FSE cusp artifact reduction when compared with two other commercially available techniques in which the slice-selection gradient polarity or the amplitude is manipulated in pulse sequence design (Figure 5.16). In addition, the slice-tilting technique can also avoid the problem with chemical shift and other off-resonance effects encountered by the commercial techniques because of the change in either gradient polarity or amplitude.

Although the primary focus of this study is not to characterize the appearance of the FSE cusp artifact, we have observed that the strength of the artifact was inversely proportional to the slice thickness. For example, the artifact strength increased from 12.2% to 19.9% when the slice thickness decreased from 8 mm to 4 mm. This dependence was most likely related to the amplitude of the slice-selection gradient. A thinner slice requires a stronger gradient, which can move the artifact-prone region closer to the sensitive region of the RF receiving coil, leading to a stronger artifact. This explanation is in agreement with the fact that the bandwidth of the RF pulse (and, thus, the slice selection gradient amplitude) can alter the appearance of the FSE cusp artifact, as exploited in one of the commercial techniques evaluated

in our study (Figure 5.16c). Interestingly, in our experimental study on human subjects, we observed no monotonic relationship between the artifact intensity and the RF excitation bandwidth. This is possibly because the continuous distribution of spins in human subjects can support a broad range of artifact-prone regions. This is different from the situation in our calibration scan where the artifact-prone region was limited to a specific location. This observation suggests that although the slice-tilting technique can be effective to avoid artifact arising from a specific location, artifact reduction may not be complete when the artifact source spans a broader region. This may explain the residual artifact after slice-tilting that we have seen in some experimental studies (e.g., Figure 5.13b, Figure 5.17b).

Another possible source for the residual artifact may arise from the location where the spatial derivative of the overall magnetic field $B_{tot} = B_0 + zG$ is zero. At this location, the slice orientation may not be effectively tilted by a tilting gradient. Since this location theoretically corresponds to only a point, the resulting residual artifact is expected to have a very narrow width, essentially turning the “feather” artifact into a thin dotted line (2-4 pixels wide), as we observed in Figure 5.17b. We have demonstrated that the simple interpolation technique can further reduce this residual artifact, as shown in Figure 5.17c. Since the column to be regenerated was very thin (typically ~ 2 -4 pixels), linear interpolation did not significantly blur the image, especially when the signal change was gradual. This interpolation algorithm was not necessary in many situations (e.g., Figure 5.14, Figure 5.15), and should be used on an “as-needed” basis either prospectively or retrospectively.

The calibration results in Figure 5.6 indicate that even though the artifact intensity was higher with a thinner slice, a smaller slice tilt was needed to create a sufficient mismatch between the slices selected by the excitation pulse and the refocusing pulses at the artifact-producing region. This geometric consideration is most likely the reason behind the linear relationship between the optimal tilt angle and the slice thickness. The two outlying points seen in Figure 5.6 are within the measurement accuracy ($\sim 0.25^\circ$), as the minimum step size in θ_y was 0.25° in the calibration. Even with the dependence on the slice thickness, the optimal tilt angle spanned only a very narrow range ($1.3 \sim 2.3^\circ$) for a slice thickness

between 3 and 6 mm. From a practical perspective, a nominal tilt angle of $\sim 2^\circ$ can be used as a default when the calibration data is not available or intentionally omitted. This can greatly simplify practical implementation of the technique.

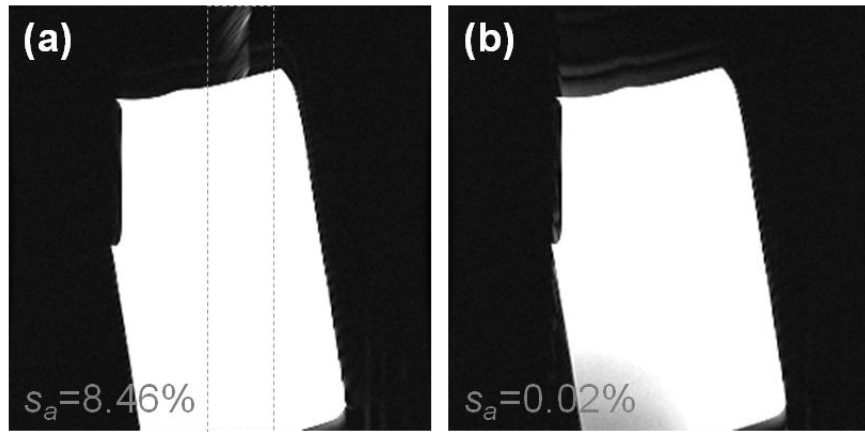


Figure 5.23: FSE images of a cylindrical phantom (a) without slice-tilting, (b) with tilt angle = 2° . Both images were obtained from a 1.5 T GE scanner quipped with a BRM (body resonance module) gradient subsystem. TR/TE = 2000/10 ms, FOV = 24 cm, slice thickness = 2 mm. Note that the ghosting artifacts near the top edge of the phantom in (a) and (b) are not the FSE cusp artifact. Instead, they are the typical FSE ghosting artifacts which we did not intend to address in this manuscript.

Although the slice tilting technique was tested on a 3.0 T and a 1.5 T scanner, both are equipped with the same gradient sub-system (CRM). The slice-tilting technique was also tested on a 1.5 T GE Signa scanner (maximum strength = 22 mT/m, maximum slew rate = 120 T/m/s) equipped with the commonly used BRM (body resonance module) gradient subsystem. The imaging parameters were the same as in the phantom experiment (see Section 5.3.1). As seen in Figure 5.23, the strength of the artifact

decreased from 8.46 % (Figure 5.23a) before the tilt to 0.02 % (Figure 5.23b) after the tilt. This experimental result suggests that the proposed technique can be extended to MRI scanners with other gradient configurations.

The proposed slice-tilting technique does have several limitations. A primary issue is the reduction in the signal away from the slice center (Figure 5.3c, Figure 5.18) because of reduced overlap between slices selected by the excitation and refocusing pulses (Figure 5.3b). However, increased gradient non-linearity (i.e., a weaker gradient) at the edge and higher B_1 -field sensitivity can make the intensity shading effect much less obvious (Figure 5.18c,d). The compensatory effect from the B_1 -field is particularly important when a phased-array coil is employed, analogous to using multi-element RF coils to offset the “center-brightening” artifacts in high-field (e.g., 3.0 T) imaging. Because of these compensatory mechanisms, the experimentally observed signal intensity near the edge of the FOV was at least 60% of the signal intensity at the center, for the FOVs and tilt angles used in this study. Further, the signal uniformity was maintained at 60-80% under most practically encountered conditions (Figure 5.18a,b). If necessary, the signal shading associated with slice tilting can be corrected for by using techniques similar to those used for surface coil intensity correction (24,131,132). Since the intensity loss can be estimated by theoretical analysis (Figure 5.3c) as well as experimental studies (Figure 5.18), algorithms for shading correction are expected to be very feasible.

Another concern on the proposed artifact reduction technique is its robustness for multi-slice imaging. With a tilted slice, inter-slice interference becomes progressively worse towards the edge of the FOV. However, with a conventional odd/even slice reordering scheme, we have observed in Figure 5.15 that the slice-tilting technique works remarkably well even without any slice gap. Figure 5.18b further shows that the signal uniformity due to slice tilting was approximately the same irrespective of single-slice or multi-slice acquisitions, indicating that slice cross-talk was insignificant in the slice-tilting technique. In the rare event that this problem becomes significant, it can be mitigated by slice re-ordering. For example, the slice order may follow a sequence such as [1, 4, 7, 10] $\rightarrow$ [2, 5, 8, 11] $\rightarrow$ [3,

6, 9, 12], instead of the typical odd/even slice ordering. If a contiguous slice ordering must be used, the inter-slice spacing can be adjusted to address this problem. Theoretically, the minimum slice spacing required for contiguous slice ordering has been determined to be:

$$sp = \tan \theta \cdot \frac{FOV}{2} + (\sec \theta - 1) \frac{th}{2} \quad [5.13]$$

where sp is the slice gap, th is slice thickness, and θ is the tilt angle. For example, when $FOV = 24$ cm, $th = 5$ mm, and $\theta = 2^\circ$, sp was estimated to be ~ 4 mm. It is worth noting that T_1 relaxation can also help recover the perturbed signals caused by tilting of the neighboring slices. Thus, the theoretical result given by Eq. 5.13 only represents the worse-case scenario and most likely will not be encountered unless contiguous slice ordering is used on spins with long T_1 relaxation times.

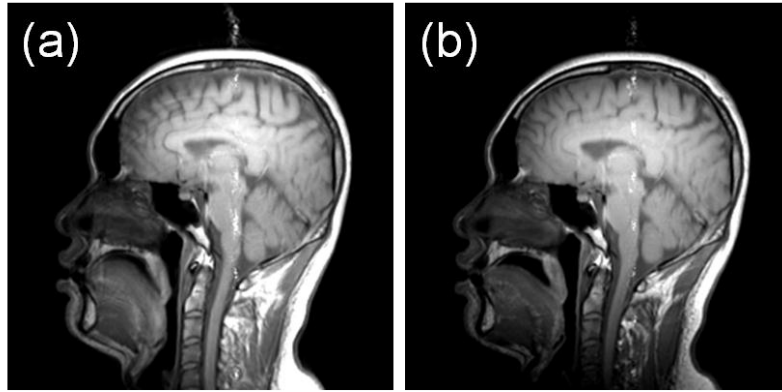


Figure 5.24: The slice-tilting approach applied to a brain image, with (a) $\theta_y = 0^\circ$, and (b) $\theta_y = 2^\circ$. The images are shown at the same window and level settings.

The third limitation of the proposed technique arises from its assumption that the artifact-

producing region resides in a defined location outside the FOV. While this assumption is valid in the vast majority of cases, physiological motion (such as respiration) can dynamically alter the artifact-producing region, which causes the artifact to have a smeared appearance and increases the challenge associated with its reduction. Figure 5.24 shows the result of applying the slice-tilting approach on brain images acquired from a healthy human volunteer. Although the artifact is shown to decrease after the tilt is applied (Figure 5.24b), it remains diffuse in appearance. This problem may be addressed by incorporating one of the motion correction or compensation techniques into slice-tilting, which is an area of further investigation.

Although all results presented in this thesis were obtained using an orthogonal tilt approach (see the Methods section), the oblique tilt approach was also implemented by including two tilting angles as described by Eqs. 5.1 and 5.5. Our experimental studies with oblique tilt indicated that the improvement over orthogonal tilt was not substantial. For example, in another experiment conducted on a phantom, the strength of the artifact decreased from 1.9% without applying any tilt angle to 0.7% with $\theta_x=0^\circ$ and $\theta_y=2^\circ$, and 0.65% with $\theta_x=\theta_y=2^\circ$. Additionally, the oblique tilt approach also exhibits a more complicated signal shading pattern. Although no significant benefit has been observed in the oblique tilt approach thus far, it does provide an additional degree of freedom to address the FSE cusp artifact and may prove useful in situations that have not been encountered in this study.

6 Phase Correction in EPI-PROPELLER

6.1 Background

PROPELLER (Periodically rotated overlapping parallel lines with enhanced reconstruction, *see* Section 4.4) (30) has been increasingly used due to its robustness against motion. Since its initial introduction based on a fast spin echo (FSE) pulse sequence (77), PROPELLER sampling schemes based on echo planar imaging (120,121) have been developed. Depending upon the direction of frequency encoding within the blade, EPI-PROPELLER has two variants: short-axis (121) and long-axis (120) PROPELLER EPI, known as SAP-EPI and LAP-EPI, respectively. Compared to FSE-PROPELLER, both EPI-PROPELLER techniques can acquire wider blades using gradient echoes, potentially improving the robustness of PROPELLER motion correction. A wider blade also allows fewer shots for the same k -space coverage. This, in conjunction with the use of a gradient echo train, can considerably reduce specific absorption rate (SAR).

All EPI-based sequences are sensitive to eddy currents due to the strong and rapidly switching acquisition gradients. Eddy currents in EPI most often result in spatially independent constant phase errors that are seen as constant phase ghosts (Figure 4.5a), and spatially linear phase errors along the readout direction that are seen as linear-phase ghosts (Figure 4.5b) on the image (*see* Sections 3.6.2 and 4.1.3) (47). In SAP- and LAP-EPI, constant and linear Nyquist ghosts can be seen in images reconstructed from each blade, resulting in a complicated “ghosting” artifact on the final PROPELLER image. Further, the changing contributions from the physical x , y , and z gradients in EPI-PROPELLER produce different phase errors for each blade, hence necessitating phase correction on a per-blade basis.

In addition to constant and linear phase ghosts in EPI-PROPELLER, a larger anisotropy between the gradient axes (32) may lead to k -space shifts in the phase encoding direction on oblique blades,

resulting in the formation of the oblique Nyquist ghost (ONG, Figure 4.5c) (see Sections 3.6.3 and 4.1.3) (33). The strength of the ONG in EPI-PROPELLER depends upon the relative anisotropy between gradient axes and the blade angle.

The cartoon in Figure 6.1 illustrates the errors in k -space that occur along the readout and phase-encoding directions as a function of the relative contributions from the physical x and y gradient axes, for three different blade orientations in EPI-PROPELLER. Due to the blade-dependent changes to the constant, linear, and oblique phase errors in EPI-PROPELLER, there exists the need to develop a robust phase-correction technique to reduce all of these commonly occurring errors simultaneously in EPI-PROPELLER.

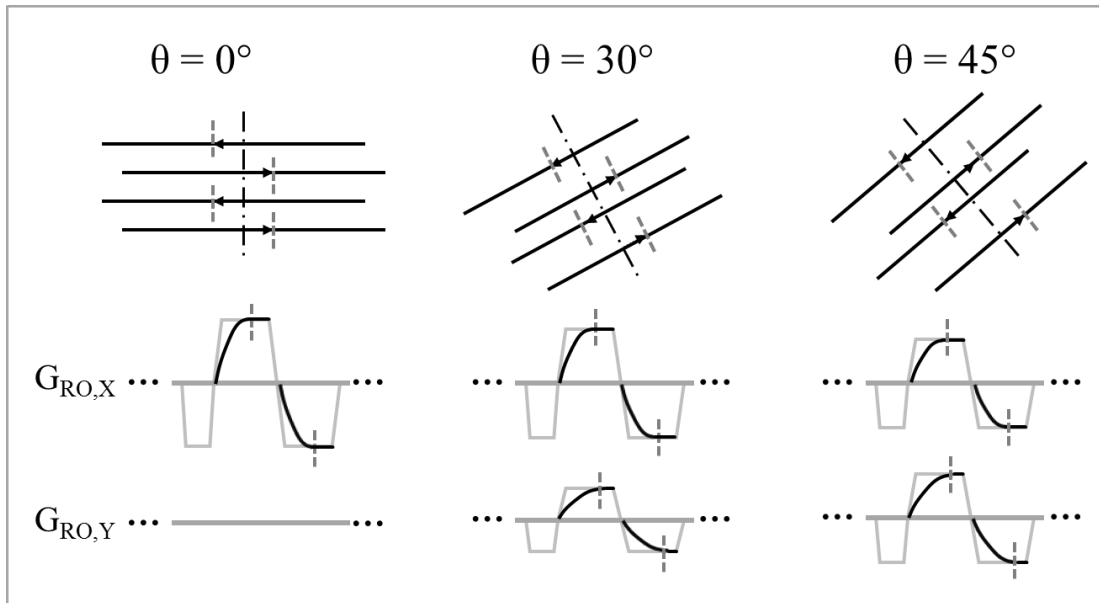


Figure 6.1: Diagram illustrating the phase errors in the readout and phase encoding directions as a function of the blade angle in EPI-PROPELLER. The phase errors are related to the gradient amplitudes along the physical x , y , and z axes.

6.1.1 *Currently Available Reduction Techniques*

Techniques that have been developed to correct for constant and linear phase errors commonly employ a reference scan (93), discussed in Section 4.1.3. However, EPI-PROPELLER requires the acquisition of blade-by-blade reference scans, which decrease the data acquisition efficiency considerably, as the duration of the scan doubles.

Several techniques that do not require a reference scan have been developed for EPI. In one technique (94), the phase errors are calculated from the point spread functions of the acquired readouts (k_x) convolved with the raw data. The phase correction algorithm involves selecting a region of interest (ROI) from a background area free of the ghost, necessitating user intervention. In another technique (133), the second moments are computed from the even and odd k -space lines, followed by phase error estimation. This technique, introduced specifically for EPI in the brain, was based on the assumption that the object was symmetric about the central axis of the image. Such an assumption does not hold for all blades in EPI-PROPELLER.

The vast majority of existing phase correction techniques focus on constant and linear phase errors. The most common technique for correcting oblique phase errors in EPI employs compensatory blip gradients along the phase-encoding direction (104). To determine the area of compensatory gradients, a calibration procedure is needed to measure the degree of gradient anisotropy. Since the oblique phase errors vary with the EPI-PROPELLER blade angle, the compensatory gradients need to be determined for each blade. Although a general scheme to obtain the phase errors was proposed (33,134), specific applications to EPI-PROPELLER have not been demonstrated. It was shown that the ONG could be further reduced by optimizing the design of the compensatory blips using a minimum entropy calculation, and is expected to be of use in EPI-based PROPELLER (135). Additionally, for real-time imaging applications requiring the acquisition of several frames for every slice, such as cardiac EPI, a double-alternating EPI sequence may be useful (136,137). This sequence was specifically designed to take advantage of the redundant temporal data acquired using parallel imaging.

Two techniques have recently been proposed for the reduction of not only the constant, linear, and oblique phase errors but also the phase errors arising from higher-order self- and cross-term eddy currents in EPI sequences. In one technique, phase errors were compensated for by using a 2D phase-map derived from phase-encoded reference scans in EPI (138). Another Nyquist ghost elimination scheme using spatial and temporal encoding (EPI-GESTE, (139)) was proposed for EPI-PROPELLER. In this technique, each line is acquired with a readout gradient of $+G_{ro}$ in one frame and $-G_{ro}$ in a neighboring frame, so that a Nyquist-ghost-free image can be reconstructed by combining data acquired in adjacent frames. Both techniques require the acquisition of additional data for each blade of EPI-PROPELLER, in the form of either phase-encoded reference scans or readouts with alternate polarity (140), leading to a decrease in data acquisition efficiency.

Although several techniques have been proposed for phase correction in EPI-based sequences, the challenges unique to EPI-PROPELLER highlight the need for a technique to simultaneously and time-efficiently perform blade-dependent correction of phase errors in the readout and phase-encoding directions, in SAP- and LAP-EPI. In this project, a generalized, time-efficient, phase correction method to simultaneously correct the most commonly seen phase errors in EPI-PROPELLER, i.e., the constant, linear, and oblique phase errors, is proposed. The technique relies on the acquisition of only two or three reference scans with the readout gradient along each orthogonal physical axis involved in producing the PROPELLER blades. For any arbitrarily oriented blade, the phase errors are obtained from these orthogonal reference scans, followed by phase corrections prior to image reconstruction of the EPI-PROPELLER images.

6.2 Phase Correction

6.2.1 Theory

We will first present our phase correction method for a generic slice orientation that involves all three physical gradient axes x , y , and z to produce a PROPELLER readout, and then reduce the formalism

to a special case of orthogonal imaging planes such as axial, coronal, and sagittal planes, which require only two physical gradient axes to generate a PROPELLER readout. For the generic case, three reference scans are acquired, each with its readout along one of the three orthogonal physical axes: x , y , and z . From each reference scan, a constant (c) and a linear (l) phase error are obtained and denoted by $(c_{||}, l_{||})$, $(c_{\perp}, l_{\perp})$, and $(c_{\odot}, l_{\odot})$ for the x -, y -, and z -axis, respectively. For an EPI-PROPELLER blade (blade angle = θ) with blade rotation matrix $\mathbf{R}_{\theta}$:

$$\mathbf{R}_{\theta} = \begin{bmatrix} \cos \theta & -\sin \theta & 0 \\ \sin \theta & \cos \theta & 0 \\ 0 & 0 & 1 \end{bmatrix}, \quad [6.1]$$

acquired in an arbitrary imaging plane defined by the rotation matrix $\mathbf{A}$:

$$\mathbf{A} = \begin{bmatrix} a_0 & a_1 & a_2 \\ a_3 & a_4 & a_5 \\ a_6 & a_7 & a_8 \end{bmatrix}, \quad [6.2]$$

the components of the logical readout (G_{ro}), phase encoding (G_{pe}) and slice selection (G_{sl}) gradients along the physical x , y , and z gradient axes are given by:

$$\begin{bmatrix} G_{ro_x} & G_{pe_x} & G_{sl_x} \\ G_{ro_y} & G_{pe_y} & G_{sl_y} \\ G_{ro_z} & G_{pe_z} & G_{sl_z} \end{bmatrix} = \begin{bmatrix} a_0 & a_1 & a_2 \\ a_3 & a_4 & a_5 \\ a_6 & a_7 & a_8 \end{bmatrix} \begin{bmatrix} \cos \theta & -\sin \theta & 0 \\ \sin \theta & \cos \theta & 0 \\ 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} G_{ro} & 0 & 0 \\ 0 & G_{pe} & 0 \\ 0 & 0 & G_{sl} \end{bmatrix} \quad [6.3]$$

$$= \begin{bmatrix} (a_0 \cos \theta + a_1 \sin \theta)G_{ro} & (a_1 \cos \theta - a_0 \sin \theta)G_{pe} & a_2 G_{sl} \\ (a_3 \cos \theta + a_4 \sin \theta)G_{ro} & (a_4 \cos \theta - a_3 \sin \theta)G_{pe} & a_5 G_{sl} \\ (a_6 \cos \theta + a_7 \sin \theta)G_{ro} & (a_7 \cos \theta - a_6 \sin \theta)G_{pe} & a_8 G_{sl} \end{bmatrix}$$

Equation 6.3 can be represented in a compact matrix form as follows:

$$\mathbf{G}_p = \mathbf{A} \mathbf{R}_{\theta} \mathbf{G}_L \quad [6.4]$$

Ignoring the contributions from the cross-term eddy currents, the x -component of the linear eddy current gradient field can be expressed by (47,92):

$$g_x(t) = \sum_i -\frac{dG_x(t)}{dt} \otimes p_{xi}(t) \quad [6.5]$$

where $G_x(t)$ is the gradient waveform along the x -axis, $p_{xi}(t)$ is a decaying exponential that characterizes a specific component (i) of the eddy currents with an amplitude and time constant, and $\otimes$ denotes convolution. With this eddy current perturbation, the k -space error along the physical x -axis is given by:

$$\Delta k'_x = \gamma \int g_x(t).dt \quad [6.6]$$

The k -space errors along the y - and z -axes can be obtained analogously.

As the amplitude of the readout gradient, G_{ro} , is much higher than that of the phase-encoding gradient, we will assume that the eddy current-related errors are predominantly caused by G_{ro} . From Eqs. 6.4–6.6, we can express the k -space shifts along the x -, y -, and z - axes as follows:

$$\Delta k'_x = \gamma G_{ro_x} \Delta t_x = \gamma G_{ro} (a_0 \cos \theta + a_1 \sin \theta) \Delta t_x \quad [6.7]$$

$$\Delta k'_y = \gamma G_{ro_y} \Delta t_y = \gamma G_{ro} (a_3 \cos \theta + a_4 \sin \theta) \Delta t_y \quad [6.8]$$

$$\text{and} \quad \Delta k'_z = \gamma G_{ro_z} \Delta t_z = \gamma G_{ro} (a_6 \cos \theta + a_7 \sin \theta) \Delta t_z \quad [6.9]$$

where Δt_x , Δt_y , and Δt_z are the time shifts of the center of the echoes due to eddy currents along the x -, y -, and z -axes, respectively, and are directly proportional to the corresponding linear phase errors.

In the presence of both linear eddy currents and gradient anisotropy, k -space shifts along the readout and phase-encoding directions are a linear combination of the errors in the physical gradient axes, and can be expressed in a manner similar to Eq. 6.4, as follows:

$$\begin{bmatrix} \Delta k'_{ro} \\ \Delta k'_{pe} \\ \Delta k'_{sl} \end{bmatrix} = (\mathbf{A}\mathbf{R}_\theta)^{-1} \begin{bmatrix} \Delta k'_x \\ \Delta k'_y \\ \Delta k'_z \end{bmatrix} = \mathbf{R}_\theta^T \mathbf{A}^T \begin{bmatrix} \Delta k'_x \\ \Delta k'_y \\ \Delta k'_z \end{bmatrix} \quad [6.10]$$

Note that $\Delta k'_{sl}$ is nullified ($\Delta k'_{sl} = 0$) for orthogonal planes, and leads to intra-slice dephasing for oblique planes if gradient anisotropy exists. In this study, our focus is on the effects of $\Delta k'_{ro}$ and $\Delta k'_{pe}$. Equation 6.10 yields the k -space shifts along the readout and the phase-encoding directions as follows:

$$\Delta k'_{ro} = \Delta k'_x(a_0 \cos \theta + a_1 \sin \theta) + \Delta k'_y(a_3 \cos \theta + a_4 \sin \theta) + \Delta k'_z(a_6 \cos \theta + a_7 \sin \theta) \quad [6.11]$$

$$\text{and} \quad \Delta k'_{pe} = \Delta k'_x(a_1 \cos \theta - a_0 \sin \theta) + \Delta k'_y(a_4 \cos \theta - a_3 \sin \theta) + \Delta k'_z(a_7 \cos \theta - a_6 \sin \theta) \quad [6.12]$$

From Eqs. 6.7–6.9 and 6.11, the generalized linear phase error in the readout direction for an arbitrary blade (with a blade rotation angle of θ) can be given as:

$$l_\theta = (a_0 \cos \theta + a_1 \sin \theta)^2 l_{\parallel} + (a_3 \cos \theta + a_4 \sin \theta)^2 l_{\perp} + (a_6 \cos \theta + a_7 \sin \theta)^2 l_{\odot} \quad [6.13]$$

Similarly, Eqs. 6.7–6.9 and 6.12 can be combined to yield the generalized oblique phase error for an arbitrary blade with an angle of θ :

$$o_\theta = \frac{FOV_{pe}}{FOV_{ro}} \frac{N_{ro}}{N_{pe}} \left\{ \frac{\sin 2\theta}{2} \{ (a_1^2 - a_0^2) l_{\parallel} + (a_4^2 - a_3^2) l_{\perp} + (a_7^2 - a_6^2) l_{\odot} \} \right. \\ \left. - \cos 2\theta \{ a_0 a_1 l_{\parallel} + a_3 a_4 l_{\perp} + a_6 a_7 l_{\odot} \} \right\} \quad [6.14]$$

where FOV_{ro} and FOV_{pe} are the fields of view in the readout and phase-encoding directions, respectively. N_{ro} and N_{pe} are the number of readout points and phase-encoding steps, respectively.

The spatially constant phase error can be produced by the gradient component along each of the three physical axes:

$$b_0(t) = - \left(\sum_{i_x} \frac{dG_x(t)}{dt} \otimes p_{0x}(t) + \sum_{i_y} \frac{dG_y(t)}{dt} \otimes p_{0y}(t) + \sum_{i_z} \frac{dG_z(t)}{dt} \otimes p_{0z}(t) \right) \quad [6.15]$$

As the primary cause of the B_0 eddy currents is the readout gradients, we can express the phase error arising from each of these three terms in a manner similar to Eqs. 6.7–6.9. Following a similar derivation as given above, the constant phase error for a blade with an angle of θ can be expressed as:

$$c_\theta = (a_0 \cos \theta + a_1 \sin \theta)c_{\parallel} + (a_3 \cos \theta + a_4 \sin \theta)c_{\perp} + (a_6 \cos \theta + a_7 \sin \theta)c_{\odot} \quad [6.16]$$

Equations 6.13, 6.14, and 6.16 describe a general approach to determine phase errors for any blade orientation from a set of orthogonal reference scans.

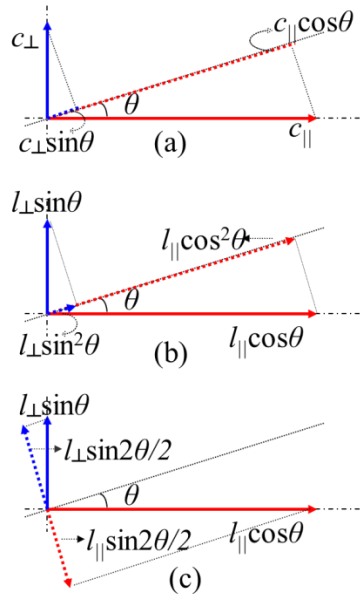


Figure 6.2: Diagrams illustrating the correction schemes for (a) constant, (b) linear, and (c) oblique Nyquist ghost in the case of an axially prescribed FOV.

In a special case where the imaging plane is normal to any of the physical gradient axes (i.e., in axial, sagittal or coronal planes), Eqs. 6.13, 6.14, and 6.16 can be considerably simplified. For example, in an axial plane where the readout gradient involves the physical x - and y -axes, Eq. 6.3 becomes:

$$\begin{bmatrix} G_{ro_x} & G_{pe_x} \\ G_{ro_y} & G_{pe_y} \end{bmatrix} = \begin{bmatrix} \cos \theta & -\sin \theta \\ \sin \theta & \cos \theta \end{bmatrix} \begin{bmatrix} G_{ro} & 0 \\ 0 & G_{pe} \end{bmatrix} \quad [6.17]$$

Following similar derivations shown above for the generic case, linear, oblique, and constant phase errors are obtained respectively as:

$$l_\theta = l_\parallel \cos^2 \theta + l_\perp \sin^2 \theta \quad [6.18]$$

$$o_\theta = \frac{FOV_{pe}}{FOV_{ro}} \cdot \frac{N_{ro}}{N_{pe}} (l_\perp - l_\parallel) \sin 2\theta \quad [6.19]$$

and
$$c_\theta = c_\parallel \cos \theta + c_\perp \sin \theta \quad [6.20]$$

In this specific case, correction of the constant, linear, and oblique phase errors for every blade can be accomplished by the acquisition of only two reference scans, illustrated graphically in Figure 6.2. The phase errors corresponding to the physical x - and y -axis are shown in red and blue respectively. From Eqs. 6.7–6.9, the linear phase error along each physical gradient axis depends upon its contribution to the readout gradient, and is effectively represented by $l_\parallel \cos \theta$ along the x -axis, and $l_\perp \sin \theta$ along the y -axis, in both, Figure 6.2b and Figure 6.2c. Constant and linear phase errors for a blade rotated by θ are shown as a linear combination of the k -space errors in the readout direction, as seen in Figure 6.2a and Figure 6.2b, respectively. The oblique phase error is a linear combination of the linear phase errors in the phase-encoding direction, shown perpendicular to the blade orientation θ , in Figure 6.2c.

Equations 6.18–6.20 are similar to those derived in (96), which described techniques to reduce the ghost on oblique EPI images by considering the orientation of the image and the contributions from the individual physical gradient axes.

6.2.2 Implementation

After data acquisition, all phase corrections were performed prior to PROPELLER image reconstruction. Figure 6.3 shows a flow chart summarizing the phase correction algorithm, which was implemented in MATLAB (The Mathworks, Natick, MA, USA). The primary steps are outlined below. First, a one-dimensional inverse Fourier transform was applied in the readout direction of every blade (f_θ), such that the blade was in the (x, k_y) space. This was followed by constant and linear phase correction using the phase errors, c_θ and l_θ , obtained from Eqs. 6.20 and 6.18 (or Eqs. 6.16 and 6.13), respectively:

$$Q_\theta(x_n, k_y) = F_\theta(x_n, k_y) e^{i(c_\theta + nl_\theta)} \quad [6.21]$$

where $Q_\theta(x_n, k_y)$ is the n th readout point in each line of the blade with an angle θ . This step is illustrated by the dotted grey lines in Figure 6.3.

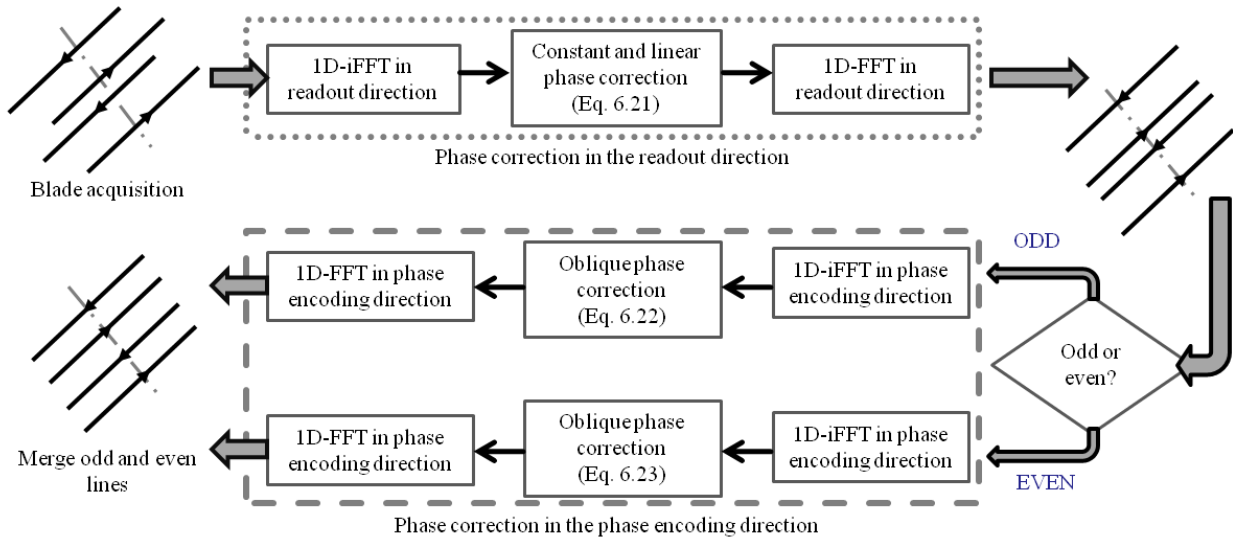


Figure 6.3: Flowchart illustrating the steps involved in blade-specific phase correction.

Following phase correction in the readout direction, the blade was transformed back into the (k_x, k_y) space, where the odd and even lines were separated to form two sub-datasets: $q_{o\theta}(k_x, k_y)$ and $q_{e\theta}(k_x, k_y)$. This was followed by an oblique phase correction using Eq. 6.19 (or Eq. 6.14), shown within the dashed grey lines in Figure 6.3. For the oblique phase correction, a one-dimensional inverse Fourier transform was applied in the phase-encoding direction on both sub-datasets, $q_{o\theta}(k_x, k_y)$ and $q_{e\theta}(k_x, k_y)$, to yield $Q_{o\theta}(k_x, y)$ and $Q_{e\theta}(k_x, y)$. The odd and even sub-dataset was then multiplied by a linear phase ramp, $+o_\theta/2$ and $-o_\theta/2$, respectively, as shown in Eqs. 6.22 and 6.23.

$$H_{o\theta}(k_x, y_m) = Q_{o\theta}(k_x, y_m) e^{im o_\theta/2} \quad [6.22]$$

$$H_{e\theta}(k_x, y_m) = Q_{e\theta}(k_x, y_m) e^{-im o_\theta/2} \quad [6.23]$$

where y_m is the m th phase-encoding step in each sub-dataset. This effectively shifted the odd k -space lines by $\Delta k'_{pe}/2$ and the even k -space lines by $-\Delta k'_{pe}/2$, removing the alternating k -space shift due to gradient anisotropy (96). The oblique error-corrected blades, $H_{e\theta}$ and $H_{o\theta}$, were transformed back into the (k_x, k_y) space and merged together, followed by PROPELLER image reconstruction.

6.3 Pulse Sequence Design

6.3.1 Implementing EPI-PROPELLER

Both SAP-EPI and LAP-EPI sequences were developed from a commercial echo planar imaging sequence and implemented on a GE Signa 3.0 T HDxt scanner (GE Healthcare, Waukesha, WI, USA) equipped with a CRM (cardiac resonator module) gradient subsystem (maximum gradient strength = 40 mT/m, maximum slew rate = 150 T/m/s). LAP- or SAP-EPI data was acquired based on the user-selected number of frequency (FR) and phase (PE) encoding steps to be acquired within each blade. The number of blades (N) was then calculated using the equation from (30), as follows:

$$N = f \frac{M}{L} \frac{\pi}{2} \quad [6.24]$$

where M is the larger of FR and PE (FR for LAP-EPI and PE for SAP-EPI),

L is the smaller of FR and PE (FR for SAP-EPI and PE for LAP-EPI), and

f is a factor indicating the degree of overlap/redundancy between blades. When $f = 1$, N is the minimum number of blades required to ensure complete k -space coverage without any overlap at the edges of the blades.

A rotation matrix ($\mathbf{R}_\theta$) with an in-plane incremental rotation angle of $\eta = \pi/N$ was applied to rotate the coordinate system for the acquisition of each blade. Phase cycling was disabled to acquire the raw data, comprising all the blades, which were saved to a single file at the end of the scan. Reference scans were acquired in two or three orthogonal directions (depending upon the selected plane) to evaluate the proposed technique. To provide a comparison of the proposed technique with the standard blade-specific phase correction technique, reference scans were also acquired in some cases for the remaining blades.

Reconstruction of the EPI-PROPELLER image comprised many steps that are common to other PROPELLER techniques, as discussed in Section 4.4.2. First, the EPI phase correction technique was implemented in MATLAB (The Mathworks, Natick, MA, USA) using Eqs. 6.21, 6.22, and 6.23. After the EPI phase correction was implemented, low-frequency phase errors were identified and removed by using a triangular window to filter them out from the high-frequency components. Next, bulk translation and rotation between blades was subtracted by using the average of the central overlapping region of all blades as a reference. The central region of each blade was correlated with the average from all the blades and this correlation factor was used to weight the information in that blade. This step removed all residual phase errors from the data. Finally, the data were gridded to Cartesian coordinates, and the image was reconstructed through a Fourier transform.

The only difference in implementing the reconstruction process for LAP-EPI and SAP-EPI was that the final image was reconstructed to a matrix size corresponding to the number of phase-encoding steps in SAP-EPI, and frequency-encoding steps in the case of LAP-EPI. The EPI-PROPELLER reconstruction algorithm was developed and implemented in the ‘C’ programming language.

6.3.2 Application to Diffusion-Weighted Imaging

The diffusion-weighted (DW) EPI-PROPELLER pulse sequence was created from a standard diffusion-weighted SE-EPI sequence. This DW SE-EPI sequence was modified in the manner described in Section 6.3.1.

In the standard PROPELLER design, the rotation matrix is used to rotate the coordinate system away from the prescribed plane for every blade, by a blade rotation angle of θ . In a diffusion imaging pulse sequence, the diffusion-weighting gradients are designed to be rotationally-invariant throughout data acquisition, i.e., the diffusion direction must remain the same regardless of the blade angle. To render the diffusion gradients rotationally invariant, a “reverse” rotation operation (i.e., rotation by an angle of $-\theta$) was performed on the diffusion-weighting gradients only, after the co-ordinate system was rotated but before the acquisition of each blade. For example, in an axial plane, the ‘reverse rotation’ or ‘un-rotation’ operation was performed as follows:

$$G_{x_{d_u}} = G_{x_d} \cos \theta + G_{y_d} \sin \theta \quad [6.25]$$

$$G_{y_{d_u}} = -G_{x_d} \sin \theta + G_{y_d} \cos \theta \quad [6.26]$$

where G_{x_d} and G_{y_d} are the diffusion-weighting gradients rotated by the blade angle θ , and $G_{x_{d_u}}$ and $G_{y_{d_u}}$ are the ‘un-rotated’ diffusion-weighting gradients applied in the pulse sequence. By performing this step, although the blade orientation changes with every excitation pulse, the diffusion direction remains consistent across the acquisition.

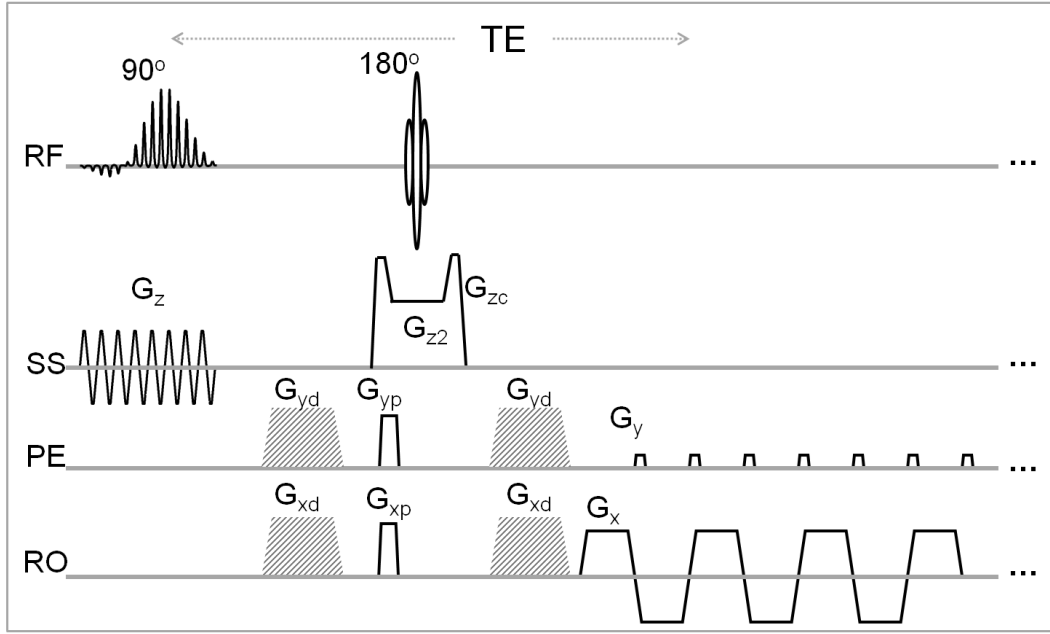


Figure 6.4: The DW-EPI-PROPELLER pulse sequence was built upon a diffusion-weighted spin-echo-based EPI pulse sequence. (The SPSP excitation pulse structure was provided by Yi Sui at the University of Illinois at Chicago.)

6.4 Methods

6.4.1 Validation Experiments

Experimental evaluation of the phase correction technique was carried out on phantoms and human subjects on the 3.0 T GE Signa HDxt scanner with a transmit-receive quadrature head coil and higher-order shims. The scope of the evaluation included T_2 -weighted imaging in orthogonal (axial, sagittal, and coronal) and oblique planes, followed by axial diffusion-weighted (DW) scans to demonstrate an application. In each evaluation, SAP- and/or LAP-EPI sequences were employed as detailed below.

To evaluate the performance of the phase correction technique in axial SAP-EPI, two orthogonal reference scans along the x - and y -axes, together with a full set of PROPELLER blades, were acquired from a square phantom (with 6 cm sides) filled with 100% acetone to reduce the dielectric resonance effect at 3.0 T. The data acquisition parameters were: TR = 4000 ms, TE = 61 ms (min. full), bandwidth = ± 62.5 kHz, acquisition matrix of 24 readout points $\times$ 128 phase-encoding steps, number of blades = 8, $f = 1$, NEX = 4, FOV = 18 cm, and slice thickness = 5 mm. For comparison, a full set of blade-specific reference scans was also acquired. Constant and linear phase errors, computed from Eqs. 6.20 and 6.18 respectively, were used with the reference scan data from the two orthogonal blades to perform phase correction on each blade (referred to as ‘Method A’ in this work), as described by Eq. 6.21. For comparison, constant and linear phase correction was independently performed using the blade-specific reference scans (referred to as ‘Method B’ in this work). In order to test not only the constant and linear phase correction steps but also the ONG correction, the first-order y -axis eddy current compensation term was altered to increase the oblique Nyquist ghost to, e.g., $\sim 10\%$ of the highest signal intensity of the image. After correcting for the constant and linear phase errors, the ONG correction was also applied, based on Eq. 6.19, and described by Eqs. 6.22 and 6.23. The image was reconstructed after phase correction. The same experiment was conducted on data acquired using LAP-EPI on the same phantom, with the following acquisition parameters: TR = 4000 ms, TE = 61 ms (min. full), bandwidth = ± 62.5 kHz, acquisition matrix of 128 readout points $\times$ 24 phase-encoding steps, number of blades = 8, $f = 1$, NEX = 4, FOV = 18 cm, and slice thickness = 5 mm.

Next, the phase correction technique was validated on sagittal and coronal planes using a spherical phantom (11 cm DSV, 3.3685 g/L $\text{NiCl}_2 \cdot 6\text{H}_2\text{O}$ and 2.4 g/L NaCl). The following acquisition parameters were used for SAP-EPI in the sagittal plane: TR = 4000 ms, TE = 55 ms, bandwidth = ± 62.5 kHz, blade matrix size = 24 readout points $\times$ 128 phase-encoding steps, number of blades = 16, $f = 2$, FOV = 20 cm, and slice thickness = 5 mm. A LAP-EPI dataset was also acquired with the same parameters, except for a transposed blade matrix. For this experiment, two reference scans were acquired with their readouts along the anterior-posterior (y -axis) and superior-inferior (z -axis) directions. For this

dataset, the first-order eddy-current compensation term in the z-axis was changed to increase the ONG. After the sagittal scan, additional experiments were carried out in a coronal plane using the same parameters.

The phase correction was further demonstrated in an oblique plane on the acetone phantom, by selecting an imaging plane rotated 45° from the axial plane, with the following common acquisition parameters: TR = 4000 ms, TE = 57 ms (min. full), bandwidth = ± 62.5 kHz, $f = 1$, NEX = 4, FOV = 22 cm, and slice thickness = 5 mm. Each LAP-EPI blade was acquired with an acquisition matrix of 128 readout points and 32 phase-encoding steps. SAP-EPI blades were acquired with an acquisition matrix of 32 readout points and 128 phase-encoding steps. Six blades were acquired in both cases. Three reference scans were acquired along the physical gradient axes, to perform constant and linear phase correction using Eqs. 6.16, 6.13, and 6.21, and oblique phase correction using Eqs. 6.14, 6.22, and 6.23.

To further demonstrate the performance of the phase correction technique in the presence of local magnetic susceptibility perturbations and with high spatial resolution, SAP- and LAP-EPI data were acquired in multiple axial planes on an ACR (American College of Radiology) phantom with internal structures. T_2 -weighted images were acquired using both sequences with the following common imaging parameters: TR = 4000 ms, TE = 84 ms, $f = 1$, NEX = 6, FOV = 42 cm, and slice thickness = 5 mm. Imaging parameters specific to LAP-EPI were: acquisition matrix per blade = 256 readout points $\times$ 16 phase-encoding steps, bandwidth = ± 125 kHz, and number of blades = 24. Parameters specific to SAP-EPI included: acquisition matrix per blade = 24 readout points $\times$ 256 phase-encoding steps, bandwidth = ± 62.5 kHz, and number of blades = 16. Similar to the experiments on the square phantom, two orthogonal reference scans were obtained, and phase corrections were performed using the flow chart in Figure 6.3.

Following the phantom validations, the phase-correction technique was applied to EPI-PROPELLER images acquired on healthy male volunteers (age: 27–31 years old). SAP-EPI images were acquired in an axial plane using the quadrature head coil with the following imaging parameters: TR =

4000 ms, TE = 96 ms, bandwidth = ± 62.5 kHz, acquisition matrix per blade = 24 readout points $\times$ 256 phase-encoding steps, number of blades = 16, $f = 1$, NEX = 4, FOV = 28 cm, and slice thickness = 5 mm. LAP-EPI data were acquired at the same location with similar imaging parameters, except for an acquisition matrix of 256 readout points $\times$ 16 phase-encoding steps for each of the 24 blades. Similar to the phantom experiments, two orthogonal reference scans were acquired and used for phase correction. To demonstrate that the phase-correction technique can be extended to oblique imaging planes, the SAP- and LAP-EPI imaging protocols were repeated (except for a larger FOV of 30 cm) in a plane along the cerebellar tentorium, rotated $\sim 40^\circ$ from the axial plane. Because all three physical gradients were involved in producing the readout gradient of the PROPELLER blade, three reference scans were acquired along each of the physical gradient axes, and Eqs. 6.16, 6.13, and 6.14 were used to correct for the constant, linear, and oblique phase errors in each blade.

In the final experiment, the phase-correction algorithm was validated on diffusion-weighted EPI-PROPELLER images (121,141). Axial short-axis (SAP) and long-axis (LAP) EPI-PROPELLER images were acquired on phantoms and a healthy volunteer, with diffusion-weighting along the principal three directions. The key data acquisition parameters for SAP-EPI included: TR = 3000 ms, TE = 130 ms (min. full), bandwidth = ± 62.5 kHz, blade acquisition matrix of 24 readout points and 256 phase-encoding steps, number of blades = 16, $f = 1$, NEX = 2, FOV = 26 cm, and slice thickness = 5 mm. Diffusion-weighted LAP-EPI was acquired with the same acquisition parameters, except for a blade acquisition matrix of 256 readout points and 24 phase-encoding steps. Diffusion-weighted images were acquired along the principal orthogonal directions, with $b = 750$ s/mm². Phase correction was performed on all images (anatomical and diffusion-weighted) using the same reference scans. For comparison, DW single-shot EPI images were also acquired with similar imaging parameters.

For all images acquired with both SAP- and LAP-EPI, an FOV greater than twice the largest dimension of the object was selected to avoid overlap between the image and the oblique Nyquist ghost, in order to facilitate illustration of the effectiveness of phase correction. Higher-order shim settings were

enabled for all datasets acquired on structured phantoms or volunteers. In all cases, an eddy-current compensation term was altered to augment the oblique Nyquist ghost and clearly illustrate its reduction.

6.4.2 Artifact Quantification

The strength of the Nyquist ghost was calculated on each blade of SAP- and LAP-EPI data before and after applying the proposed phase correction steps. Three regions of interest (ROI), each comprising ~ 100 pixels, were selected from within the images obtained from every blade, as follows: (a) in a homogeneous region of the object with the highest signal intensity (for the head images from the human volunteer, the ROI was placed within the ventricles as it had the most uniform signal intensity), (b) in the background (air) without the artifact, and (c) in the background containing the strongest artifact intensity. The intensity of the Nyquist ghost (g_N) was quantified by the following equation:

$$g_N = \frac{\mu_g - \mu_n}{(\mu_s - \mu_n) + (\mu_g - \mu_n)} \quad [6.27]$$

where μ_g was the mean signal intensity of the ROI over the artifact, μ_s was the mean signal intensity of the ROI over the object, and μ_n was the mean signal intensity of the ROI over the background.

After image reconstruction, the Nyquist ghost is smeared throughout the background of the reconstructed EPI-PROPELLER image. It is difficult to quantify these non-localized ghost artifacts using standard ghost-quantification techniques. As the ROI-based artifact evaluation method discussed above may not be capable of quantifying the artifact on the reconstructed image accurately, the average ghost per blade was quantified using Eq. 6.27, and was used as a measure of the performance of the phase-correction in all studies.

6.5 Results

The performance of the phase correction technique is shown on a SAP-EPI dataset in Figure 6.5. Figure 6.5a shows individual blades of the dataset without any phase correction applied. The result of

correcting the constant and linear phase errors from the synthesized reference datasets ('Method A') is shown in Figure 6.5b. The constant and linear ghosts decrease from an average of 38.2% without the phase correction to 12.7% per blade after the phase correction. (The residual ghosts were related to the ONG and will be discussed shortly). This technique is comparable to using blade-specific reference scans ('Method B'), the results of which are illustrated by Figure 6.5c, with an average ghost intensity of 12.8% per blade. After the oblique phase correction was applied to the data corrected using Method A, the ghost further decreased to 0.7% for the non-orthogonal blades, as shown in Figure 6.5d. Figure 6.5a-d are shown with the window and level settings adjusted to highlight the background. Figure 6.5e-h show the images that were reconstructed from Figure 6.5a-d, respectively, at the same window and level settings as Figure 6.5a-d. In the PROPELLER image without correction (Figure 6.5e), the Nyquist ghost artifacts were delocalized and smeared with a reduced intensity, giving a shadow-like "polygon" appearance. After constant and linear phase correction using either the proposed method A (Figure 6.5f) or the blade-by-blade Method B (Figure 6.5g), the artifacts were considerably reduced. The equivalence of the results in Figure 6.5f and Figure 6.5g indicated that the proposed time-efficient phase correction technique performed at least equally well when compared to the time-consuming method employing considerably more reference scans. Figure 6.5h shows further reduction of the smeared ghost artifacts in the background after correction of the oblique phase errors. Figure 6.5e'-h' show the same image under standard window and level settings, such that the shading across the object (Figure 6.5e') is seen to progressively decrease after (Figure 6.5f'-h') the phase correction steps were applied.

Similar results can be seen from the LAP-EPI dataset (Figure 6.6), demonstrating a reduction of the ghost from 40.3% before the phase correction (Figure 6.6a) to 12.2% after the constant and linear phase correction using Method A (Figure 6.6b) and 12.1% using Method B (Figure 6.6c), and further to 1.4% after the ONG correction technique is applied (Figure 6.6d). Window and level settings have been adjusted in Figure 6.6a-d to highlight the background. Figure 6.6e-h demonstrate the reduction in the ghost on reconstructed images. The ghost appears as a smearing of the signal in the background, and is

reduced after application of Methods A (Figure 6.6f) and B (Figure 6.6g). Figure 6.6f and Figure 6.6g demonstrate the overall similarity of the proposed phase correction technique with the standard method employing blade-specific reference scans. The ONG correction further reduces the smearing in the final image (Figure 6.6h). Figure 6.6e'-h' show the corresponding reconstructed images from Figure 6.6e-h at standard window and level settings.

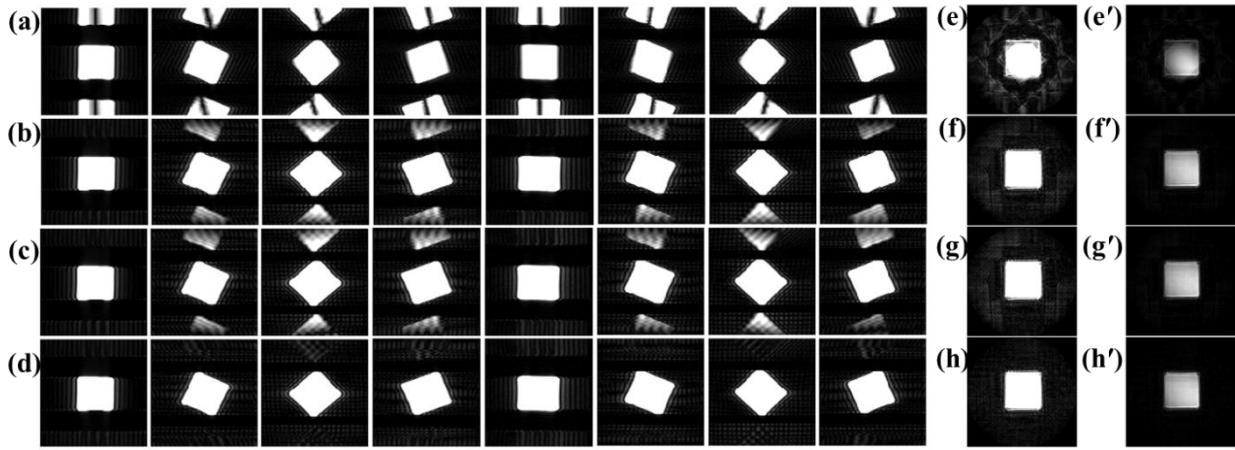


Figure 6.5: The performance of the phase correction technique on SAP-EPI images from the acetone phantom is illustrated. The panels (a-d) show the images from individual blades without any phase correction (a, $g_N = 38.2\%$), after the constant and linear phase correction step has been applied using Method A (b, $g_N = 12.7\%$) and Method B (c, $g_N = 12.8\%$), and after correction for ONG (d, $g_N = 0.7\%$). The window and level settings have been adjusted in (a-d) to display the background. The panels in e-h, and e'-h' show the reconstructed images from (a-d) respectively, at the adjusted window and level settings (e-h), and at standard window and level settings (e'-h').

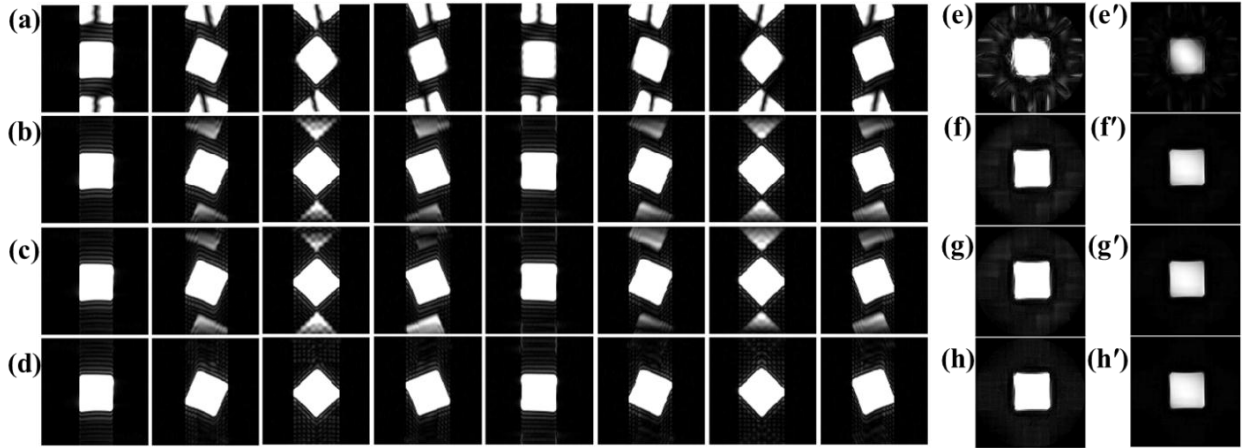


Figure 6.6: The performance of the phase correction technique on LAP-EPI images from the acetone phantom is illustrated. The panels (a-d) show the images from individual blades without any phase correction (a, $g_N = 40.3\%$), after the constant and linear phase correction step has been applied using Method A (b, $g_N = 12.2\%$) and Method B (c, $g_N = 12.1\%$), and after correction for ONG (d, $g_N = 1.4\%$). The window and level settings have been adjusted in (a-d) to display the background. The panels in e-h, and e'-h' show the reconstructed images from (a-d) respectively, at the adjusted window and level settings (e-h), and at standard window and level settings (e'-h').

A representative result of applying the phase-correction technique on a LAP-EPI image acquired on a sagittal plane is shown in Figure 6.7. Figure 6.7a shows the change in the angle of the null, establishing the presence of the ONG and indicating its variation with the blade angle. The ghost decreased from 41.3% before (Figure 6.7a) to 16.7% after (Figure 6.7b) constant and linear phase correction; and further to 3.4% (Figure 6.7c) after applying the ONG correction in the phase-encoding direction. Reconstructed images corresponding to (Figure 6.7a–c) are shown in (Figure 6.7a'–c'), and clearly illustrate decreased smearing in the background due to phase correction. The absence of any significant geometric distortion or signal intensity non-uniformity on Figure 6.7c' indicates that, with

phase correction, EPI-PROPELLER can be used to acquire good-quality images on non-axial planes. Similar results of the phase correction were obtained on sagittal SAP-EPI images, where the ghost decreased from 21.2% before correction to 13.4% after constant and linear phase correction, and further to 0.9% after ONG correction (images not shown). Comparable results were obtained in the coronal planes for both SAP- and LAP-EPI.

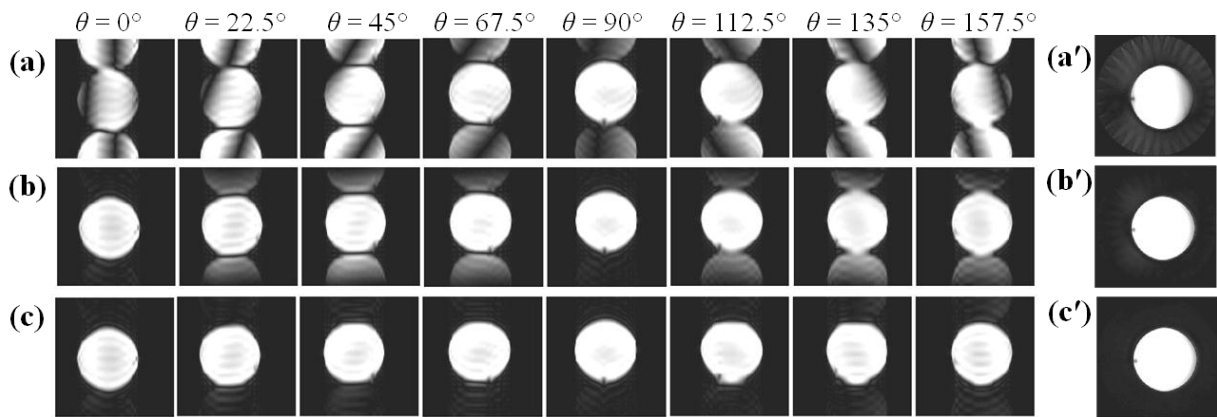


Figure 6.7: The phase correction has been evaluated on sagittal LAP-EPI images on the spherical phantom. The panels (a-c) show the images from individual blades without any phase correction (a, $g_N = 41.3\%$), after the constant and linear phase correction step has been applied (b, $g_N = 16.7\%$), and after correction for ONG (c, $g_N = 3.4\%$). The window and level settings have been adjusted in (a-c) to display the background. The panels in (a'-c') show the reconstructed images from (a-c) respectively, at the adjusted window and level settings.

The results of applying the proposed phase-correction algorithm on the square phantom in the oblique plane are illustrated by the SAP-EPI images in Figure 6.8. The average ghost per blade decreases

from 26.0% (Figure 6.8a) to 5.7% (Figure 6.8b) after constant and linear phase correction is performed on SAP images. After phase correction in the phase-encoding direction, the average ghost per blade decreases even more to 1.4% (Figure 6.8c). The panel on the right (Figure 6.8a'–c') shows the reconstructed SAP images. Similar results were obtained in LAP-EPI, where the average ghost per blade decreased from 33.9% before to 10.9% after correction for constant and linear phase errors, and further to 2.3% after ONG correction (images not shown). In both cases, constant and linear phase corrections improve the quality of the image substantially, whereas the smeared ghost in the background is reduced after the ONG is corrected.

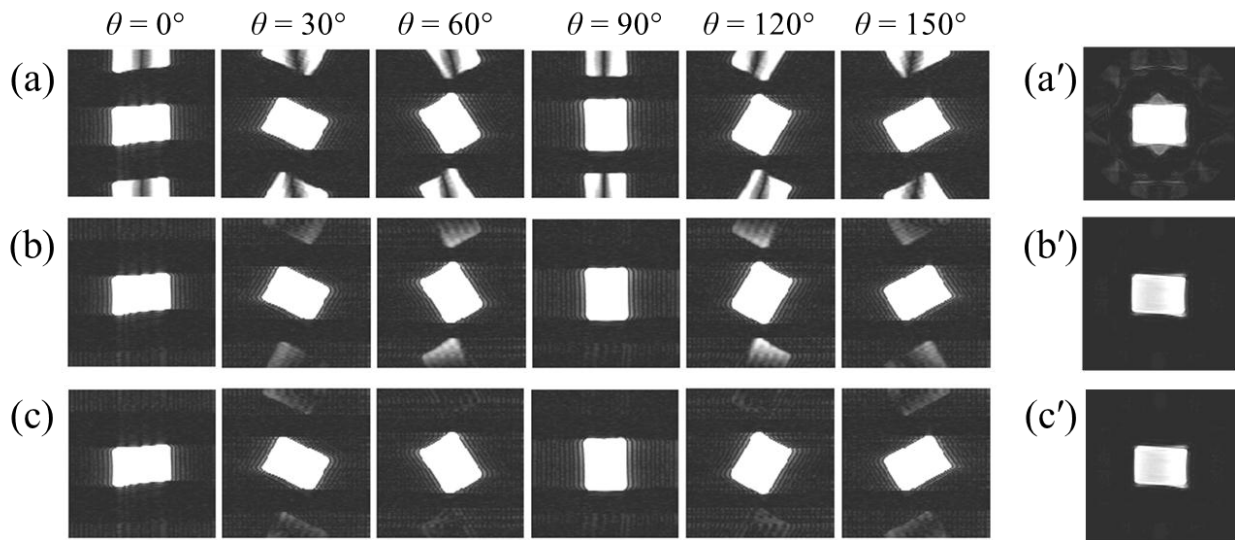


Figure 6.8: The phase correction evaluated on SAP-EPI images from the acetone phantom is illustrated. The panels (a-c) show the images from individual blades without any phase correction (a, $g_N = 26.0\%$), after the constant and linear phase correction step has been applied (b, $g_N = 5.7\%$), and after correction for ONG (c, $g_N = 1.4\%$). The window and level settings have been adjusted in (a-c) to display the background. The panels in (a'–c') show the reconstructed images from (a-c) respectively, at the adjusted window and level settings.

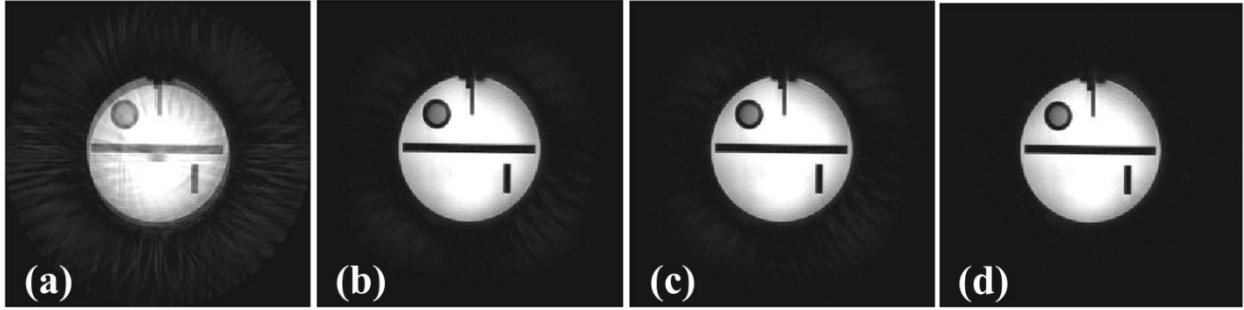


Figure 6.9: Reconstructed LAP-EPI images of the ACR phantom, (a) without any phase correction, and after constant and linear phase correction using Method A (b) and Method B (c). The Nyquist ghost decreased from an average of $g_N = 48.9\%$ for each blade before phase correction to $g_N = 19.1\%$ after applying Methods A and B, and $g_N = 4.6\%$ after further correction for the ONG (d).

Figure 6.9 illustrates the performance of the phase correction technique on one of the axial images acquired from the ACR phantom using LAP-EPI. Figure 6.9a shows the image reconstructed from data without any phase correction. The ghost decreased from an average of 48.9% (Figure 6.9a) per blade before to 19.1% after the constant and linear phase correction techniques using both Methods, A (Figure 6.9b) and B (Figure 6.9c). Further reduction of the ONG was achieved, to 4.6% (Figure 6.9d). A similar performance was noticed on another axial slice acquired from this phantom (Figure 6.10) using SAP-EPI. In this case, the ghost decreased from 40.5% before phase correction was applied (Figure 6.10a) to 2.8% after correcting for constant, linear and oblique phase errors (Figure 6.10b). Figure 6.10 is shown with the window and level settings adjusted to highlight the background. The blurring seen at the top of the phantom is the result of susceptibility differences at the fluid-air interface.

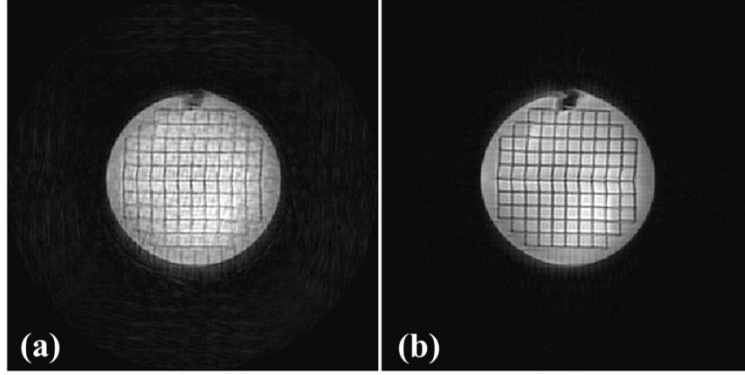


Figure 6.10: Reconstructed SAP-EPI images of the ACR phantom, (a) without any phase correction, and (b) after correcting for constant, linear and oblique phase errors using the proposed method. The Nyquist ghost decreased from $g_N = 40.5\%$ for each blade before to $g_N = 2.8\%$ after the correction was applied. The window and level settings have been adjusted to highlight the background.

Figure 6.11 demonstrates the performance of the phase correction technique applied to the human volunteer. The Nyquist ghost seen on the axial images reduced from 26.7% for SAP-EPI and 38.6% for LAP-EPI for every blade before (Figure 6.11a,c) to 0.8% for SAP and 4.8% for LAP (Figure 6.11b,d) after correcting for constant, linear, and oblique phase errors on each blade. This result is similar to the result of applying the phase correction technique evaluated on an arbitrarily selected non-axial plane from the volunteer for both SAP (third column) and LAP (fourth column). The ghost decreased from 31.3% and 36.5% (Figure 6.11e,g) to 2.7% and 5.0% (Figure 6.11f,h) for SAP (Figure 6.11e,f) and LAP (Figure 6.11g,h), respectively. In addition, the non-uniformity on the reconstructed images was also greatly reduced. Similar results were obtained on the second volunteer and on other oblique planes (data not shown).

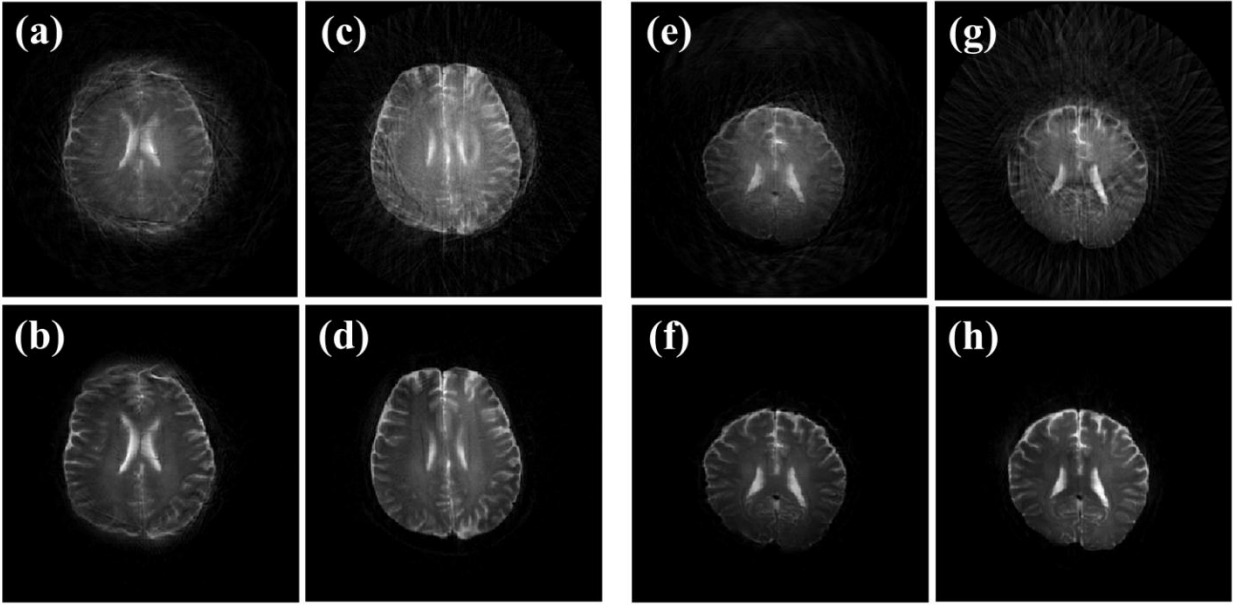


Figure 6.11: The phase-correction technique is illustrated on EPI-PROPELLER images acquired on axial (a-d) and oblique (e-h) planes. (a-d) show SAP-EPI (a,b) and LAP-EPI (c,d) images before (a,c) and after (b,d) phase correction using Eqs. 6.18-6.20 on data acquired in an axial plane. (e-h) show SAP-EPI (e,f) and LAP-EPI (g,h) images before (e,g) and after (f,h) phase correction using Eqs. 6.13, 6.14, and 6.16 on data acquired in an oblique plane, along the cerebellar tentorium.

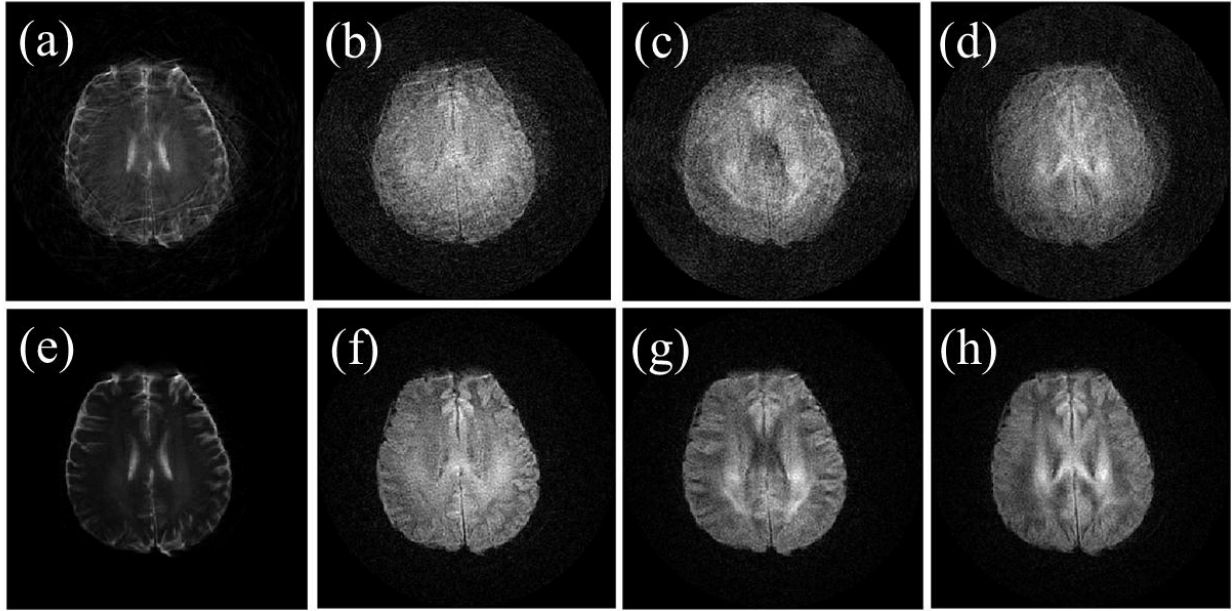


Figure 6.12: (a, e) Axial T_2 -weighted images acquired using SAP-EPI, with diffusion-weighting ($b = 750$ s/mm²) in the superior-inferior (b, f), right-left (c, g) and anterior-posterior (d, h) directions, before (a–d) and after (e–h) phase correction.

Figure 6.12 and Figure 6.13 show the results of applying the phase correction to diffusion-weighted images in the axial plane. Figure 6.12a and Figure 6.12b show SAP images (Figure 6.13a and Figure 6.13b for LAP) before and after the phase-correction, respectively; these results are similar to those in Figure 6.11a–d. SAP (Figure 6.12b–d) and LAP (Figure 6.13b–d) images acquired with diffusion weighting along the superior-inferior, left-right and anterior-posterior directions, respectively, were corrected using the same reference scans as the anatomical images and showed considerable improvement after the phase correction steps were applied (Figure 6.12f–h and Figure 6.13f–h). Diffusion-weighted single-shot EPI images (Figure 6.14) provide a visual reference to the SAP-EPI images. The spuriously high signal intensities (signal ‘pile-up’) in the frontal brain on the single-shot EPI images are due to local field inhomogeneities, and can be avoided by using EPI-PROPELLER.

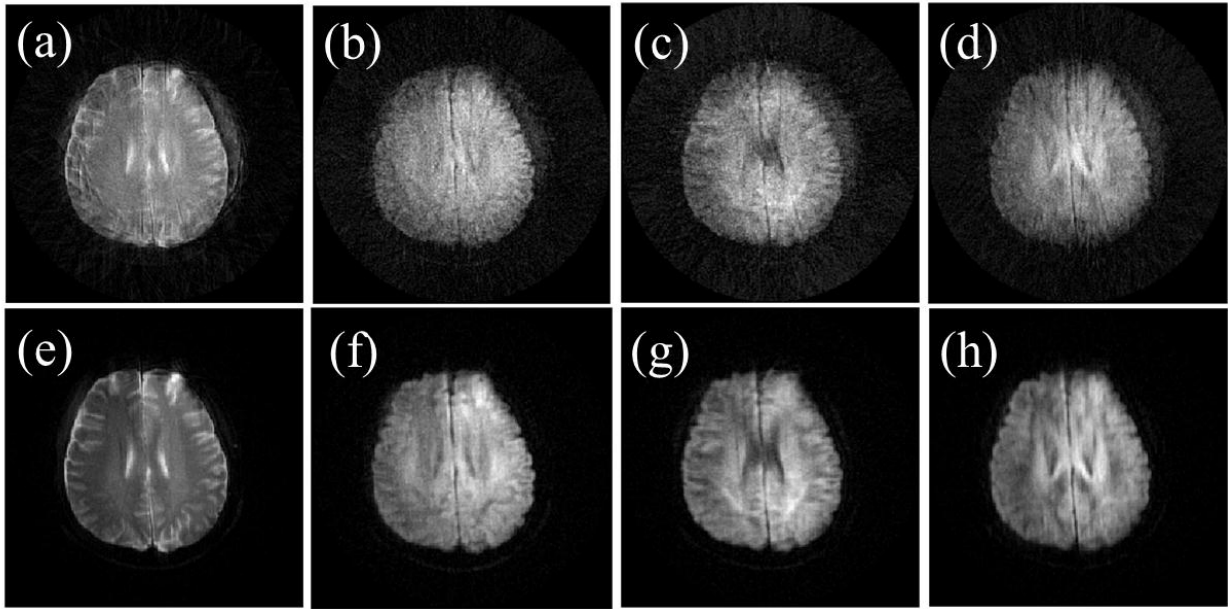


Figure 6.13: (a, e) Axial T_2 -weighted images acquired using LAP-EPI, with diffusion-weighting ($b = 750$ s/mm²) in the superior-inferior (b, f), right-left (c, g) and anterior-posterior (d, h) directions, before (a–d) and after (e–h) phase correction.

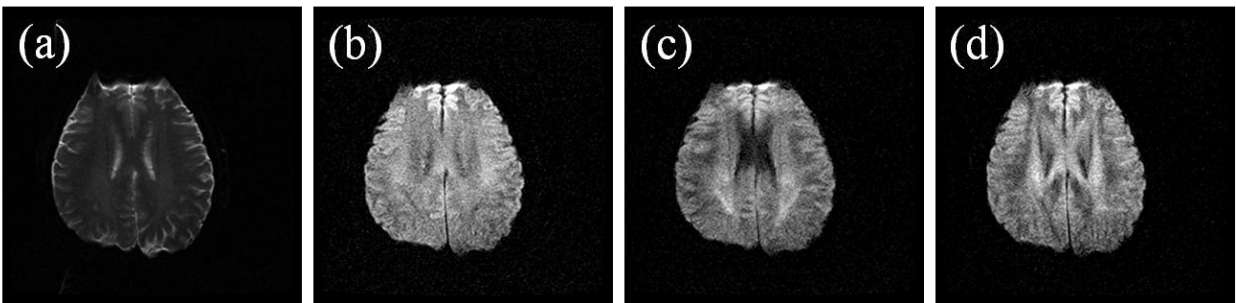


Figure 6.14: Single-shot EPI images, with (a) T_2 -weighting, and (b–d) diffusion weighting ($b = 750$ s/mm²), along the superior-inferior (b), right-left (c), and anterior-posterior (d) directions.

6.6 Discussion

We have developed a phase correction technique for EPI-PROPELLER to simultaneously, time-efficiently, and effectively reduce constant, linear, and oblique phase errors. A major novelty of the proposed technique lies in its ability to address the three types of phase errors in a simultaneous and unified fashion. The method uses only two or three reference scans, irrespective of the total number of blades, the specific blade orientations, or the prescribed imaging planes, which offers a considerable time-saving as compared to blade-by-blade reference scans. The phase correction technique has been demonstrated in both SAP-and LAP-EPI sequences on phantoms and human subjects. Excellent artifact reduction was observed in both orthogonal and oblique planes.

The proposed phase correction technique has proven to be remarkably effective in all the experiments we have conducted. As seen in Figure 6.5 and Figure 6.6, the phase correction technique reduced constant and linear ghosts in the individual blades from ~39% to ~12%, and further to ~1% after the ONG correction. Similar ghost reduction with a final ghost level of 0.5~5% was observed in all other experiments involving both orthogonal and oblique planes, low and high resolution images, SAP- and LAP-EPI sequences, phantoms and human volunteers, and T_2 -weighted and DW contrasts.

The most commonly used method to reduce constant and linear ghosts in EPI-based sequences relies on the acquisition of reference scans for each acquisition. As reference scans are acquired for every dataset, they can accurately determine the phase errors in each EPI acquisition, which are then used for phase correction. The superior phase correction performance is accompanied by a penalty in time-efficiency in EPI-PROPELLER where blade-specific reference scans are needed. Blade-specific reference scans also cannot correct for the ONG seen in non-orthogonal blades. The proposed phase correction technique maintains the flexibility and performance of reference-scan based phase correction while significantly increasing its time-efficiency. For example, the proposed technique has resulted in a 4-fold reduction of the time required for reference scanning when the total number of blades acquired was 8 (illustrated by the experiments with the acetone phantom). Considerably more scan time reduction occurs

with an increased number of blades, e.g., for the LAP-EPI axial studies on the volunteer, the duration of the reference scan was reduced from 96 s to 8 s, i.e, a 12-fold reduction. Further, in an application such as diffusion imaging, all of the acquired images can be phase-corrected with the same set of reference scans, resulting in even more improvements in efficiency. Apart from phase correction in the readout direction, this approach can also use the non-phase-encoded reference scans for oblique ghost reduction in the phase-encoding direction. This technique can also be extended to multi-slice acquisitions, as well as applications such as diffusion-weighted imaging, with no further increase in the time needed to acquire reference scans.

The majority of EPI phase correction techniques focus on constant and linear phase errors without consideration of oblique phase errors (93,94,133). When the constant and linear phase errors are the primary causes of Nyquist ghosts, our technique performs at least equally well as compared to the more time-consuming method that acquires a full set of reference scans, as shown in Figure 6.5 and Figure 6.6 (second vs. third row) for both SAP- and LAP-EPI. When the oblique phase errors are also present, the proposed technique can accurately estimate these phase errors using the existing reference scans, thus unifying the correction of all three types of Nyquist ghosts. This ability is particularly relevant to EPI-PROPELLER where oblique phase errors can be important because of the rotating readout gradient.

The oblique phase errors in EPI-PROPELLER arise primarily from gradient anisotropy (32,33,96,103,104,134), although cross-term eddy currents from the readout axis (donor) to the phase-encoding axis (recipient) can also contribute (47). In this study, our effort was concentrated primarily on gradient anisotropy as a source of oblique phase errors. Although the 3.0 T scanner employed in this study was well calibrated for gradient anisotropy, in general residual gradient anisotropy is often observed in commercial scanners. Even for a well-calibrated scanner, degradation over time can occur, leading to worsening ONGs. Although the oblique phase errors may not be as prevalent as the constant and linear phase errors, they can degrade EPI-PROPELLER images as shown in Figure 6.5–Figure 6.8. The phase correction based on Eqs. 6.14 and 6.19 can provide an effective defense against oblique phase error in

EPI-PROPELLER. In addition, Eqs. 6.14 and 6.19 can also serve as a diagnostic tool to quantify the residual gradient anisotropy in an MRI system.

To reduce oblique phase errors (or k -space shifts) in the phase-encoding direction, other robust correction techniques were recently proposed (138-140). These techniques rely on additional data in the form of phase-encoded reference scans (138) or alternating readout gradient polarity (139), with data acquisitions performed on a per-blade basis when adapted to EPI-PROPELLER (140). The blade-by-blade reference data acquisition can compromise the overall data acquisition efficiency, as further discussed shortly. A major advantage of these approaches, however, is that they can address phase errors due to higher-order eddy currents. Fortunately, for the majority of commercial MRI scanners, higher-order eddy currents are insignificant and do not lead to substantial artifacts. The inability to correct for higher-order eddy currents in our proposed method is not expected to be a major limitation.

It is interesting to note that the distinctive, well defined Nyquist ghosts in the individual blades became delocalized and diffused in the final PROPELLER image (Figure 6.5e, Figure 6.6e, Figure 6.7a', and Figure 6.8a'). This ghost transformation is a reflection of the hybrid nature of PROPELLER k -space trajectories between rectilinear and radial sampling schemes. As well studied, aliasing artifacts (such as the Nyquist ghosts) give rise to the “radiating spokes” and image blurring in radial k -space imaging, instead of localized ghosts as in rectilinear k -space imaging (142,143). The signatures of the radial k -space aliasing artifacts can be seen in the background of Figure 6.5e–g, Figure 6.6e–g, Figure 6.7a', b', and Figure 6.8a', b'. Concurrent to the ghost transformation from the individual blades to the PROPELLER image, the Nyquist ghost intensity was also considerably attenuated because the energy is spread more evenly in the background. This can be a desirable property of PROPELLER imaging, which transforms a ghost level of ~5% in the individual blades to virtually undetectable in the final PROPELLER image. Because of the diffused nature of the Nyquist ghosts and a significant ghost intensity reduction in the final PROPELLER image, we chose to quantify the ghost reduction efficiency

based on individual blade images (Eq. 6.27), which represents a more stringent measure for ghost evaluation.

As suboptimal B_0 field uniformity can affect each blade in EPI-PROPELLER differently, causing blade-specific image distortion and interfering with the subsequent reconstruction, all volunteer and structured phantom images shown were acquired by enabling higher-order shim settings. Figure 6.15 shows a SAP-EPI image acquired by disabling (Figure 6.15a) and enabling (Figure 6.15b) the higher-order shim. The shading on the lower edge of the phantom illustrates the necessity for higher-order shim; however, it can be seen that the phase correction is equally effective even in the presence of B_0 or B_1 inhomogeneity.

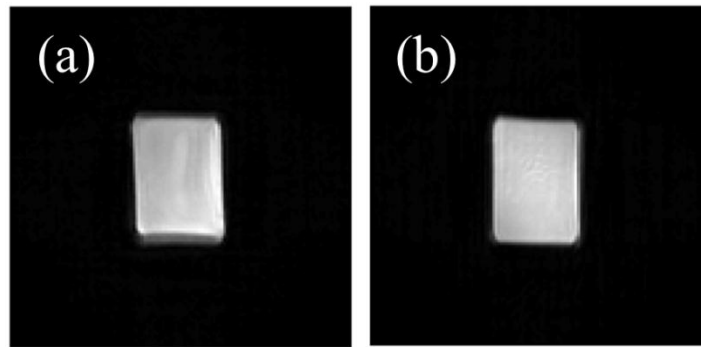


Figure 6.15: SAP-EPI images after phase-correction, (a) disabling higher-order shim, and (b) with higher-order shim enabled.

For all studies, the FOV was typically selected to be approximately twice the largest dimension of the object. This was done not only to demonstrate the reduction of the Nyquist ghost adequately, but also to avoid overlap between the ghost and the object. Using a larger FOV was especially important in the

context of oblique ghost correction, when the odd and even lines of k -space are separated, effectively decreasing the image FOV by half. Any smaller phase discontinuities that remain or are introduced after the ONG correction can compromise the reconstructed images. The object to FOV ratio is more important when the object contains sharp edges that introduce phase discontinuities. For example, the ACR images were affected by the use of a 30-cm FOV, as seen in Figure 6.16. Additionally, the choice of FOV was more important when SAP (Figure 6.16a) rather than LAP (Figure 6.16b) was used, due to the already significant presence of Gibbs ringing due to the smaller readout in SAP-EPI acquisitions. However, volunteer images were observed to be less influenced by the choice of FOV. This phase-correction technique was not evaluated for tight or reduced FOVs (i.e., in situations where the object dimension was larger than the selected FOV), where multiple copies of the ghost can overlap with the object.

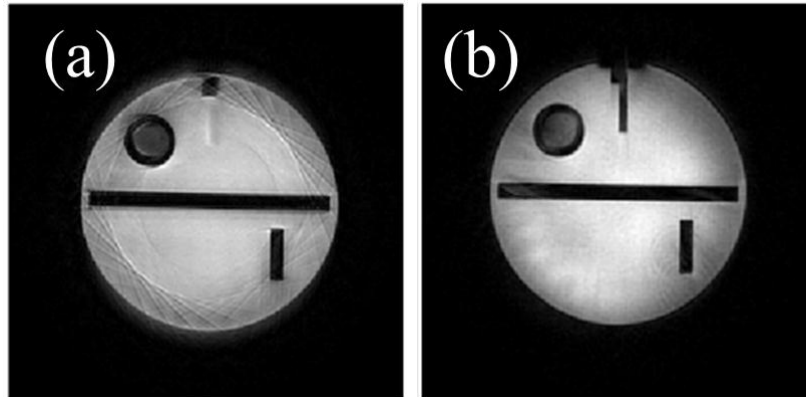


Figure 6.16: (a) SAP-EPI and (b) LAP-EPI acquired with FOV = 30 cm, and corrected for constant, linear, and oblique phase errors.

One of the possible concerns with a reference scan-based method for phase correction is the dependence of the calculated phase errors on factors such as chemical shift and susceptibility effects introduced by the object to be imaged. If the phase errors calculated along the orthogonal gradient axes are biased by object-dependent phase, this bias can propagate onto the synthesized blade-specific phase-errors, and potentially affect the efficiency of phase correction in the non-orthogonal blades. Although these factors were not seen to impact the performance of the phase correction in our images, our technique can alternately use the phase errors calculated from a balanced reference scan (144) to reduce the object-dependent phase. This step will increase the duration of the scan by ~ 2 -3 excitations (i.e., the time needed to acquire two or three additional reference scans with opposite gradient polarity), but the time penalty will still be lower than if blade-specific reference scans had been acquired.

Another source of phase error arises from concomitant field gradient terms, which are more pronounced in non-axial planes, and may compromise images acquired using large readout gradients (145-149). Concomitant field gradients have a quadratic spatial dependence which can have significant effects in arbitrary oblique planes. The proposed phase correction, which is based on a conventionally acquired reference scan, is unable to correct for the errors due to Maxwell terms, which are dependent on the spatial location in the image as well as in k -space. These phase errors decrease with increasing B_0 -field, and are estimated not to be significant in this study performed at 3.0 T. At a lower field where the problem is more relevant, the phase errors from concomitant magnetic fields can be computed accurately if the imaging parameters for an experiment are known. The Maxwell correction is therefore independent of the proposed phase correction.

Although this chapter has focused almost exclusively on the linear phase errors in readout and phase-encoding direction, Eq. 6.10 shows that there can also exist a linear phase error in the slice-selection direction, given by:

$$\Delta k'_{sl} = a_2 \Delta k'_x + a_5 \Delta k'_y + a_8 \Delta k'_z \quad [6.28]$$

where $\Delta k'_x$, $\Delta k'_y$, and $\Delta k'_z$ are given by Eqs. 6.7–6.9, respectively. From these equations and Eq. 6.28, we

can infer that the linear phase error in the slice-selection direction only exists for true oblique planes and in the presence of significant gradient anisotropy. The result of this linear phase dispersion in the slice profile is a global loss of signal across the slice. It is very likely that, in EPI-PROPELLER, the signal loss occurring due to $\Delta k'_{sl}$ can be compensated for by the increased signal from oversampling the center of k -space.

7 Conclusions

7.1 Summary

The goal of this PhD project was to demonstrate techniques for the reduction of two artifacts, both of which arise as a result of imperfections in the gradient field in MRI. The first artifact is the cusp artifact, which has been observed predominantly as a feather or streak on FSE images acquired in sagittal and coronal planes. The cusp artifact is an aliasing artifact that arises due to gradient non-linearity at the edge of the prescribed FOV. We have demonstrated that the introduction of a small tilt in slice selection during the RF excitation pulse can decrease the FSE cusp artifact by at least 50% on phantom images, and more than 90% on images acquired on the human thoracic spine, where the artifact is most problematic. This approach involves only a simple modification to the pulse sequence together with an easy calibration procedure without any change to the hardware, patient handling, or image reconstruction algorithms. The image is produced in final or near-final form at the end of the scan, and can be further improved with a straightforward post-processing step. This simple and effective approach can be applied to virtually all MRI scanners in which the FSE cusp artifact is problematic.

The second artifact is the Nyquist ghost artifact seen on EPI-PROPELLER images. This artifact originates from the eddy currents induced due to the gradient fields in EPI-PROPELLER, and forms a complicated ghosting pattern that can overlap with the object and obscure its features. Reference scans acquired in two or three orthogonal planes can be used to calculate the phase errors for any arbitrary blade in EPI-PROPELLER. These phase errors can be used to perform blade-specific phase-correction to reduce the constant- and linear-phase ghosts along the readout direction with a performance comparable to the use of blade-specific reference scans, with improved time-efficiency. Additionally, a method is proposed to reduce the oblique Nyquist ghost along the phase-encoding direction in EPI-PROPELLER.

The combined algorithm provides a way to correct for the most commonly-seen ghost artifacts due to induced eddy currents in EPI-based PROPELLER. The application of this technique to several phantoms and human volunteers has demonstrated a reduction in the constant and linear phase Nyquist ghost to as little as 2~3 % when gradient anisotropy was not significant. Even in the presence of anisotropy, the Nyquist ghost can be reduced to 0.5~5 % after the oblique phase correction step. Further, the phase correction can be applied to diffusion-weighted LAP- and SAP-EPI images, with similar results. Although the technique was developed specifically for EPI-PROPELLER, it can be applied to any EPI-based acquisition sequence in any arbitrary orientation.

7.2 Contributions

This PhD project focused on the development and validation of two techniques for the reduction of artifacts seen in fast MRI.

In *Aim 1*, a pulse-sequence-based slice-tilting approach (150) was proposed and validated to reduce the FSE cusp artifact encountered in clinical imaging. This pulse-sequence design is different from standard inner-volume imaging techniques (34,35), where the excited slice is chosen orthogonal to the refocused slice to get a non-aliased MR image from a reduced FOV. Although variants of this technique employing a large tilt angle and a thicker slice have been developed, e.g., ZOOM-EPI (130), cross-talk in multi-slice acquisitions provides a significant challenge to the use of such techniques. In contrast, the proposed slice-tilting technique uses a tilt angle that is small ($1\sim3^\circ$) and optimally selected to avoid exciting signal from the gradient null region that causes the cusp artifact. Selecting a smaller tilt angle allows the slice thickness and slice selection profile to be maintained without significantly compromising the signal within a larger FOV. The slice tilting approach was also successfully applied toward the reduction of aliasing artifacts when non-axial slices were selected in FSE-based PROPELLER (151).

In *Aim 2*, a technique was proposed to reduce the Nyquist ghosts in EPI-PROPELLER. The phase errors obtained from two or three orthogonal reference scans were used to synthesize and correct the phase errors in the remaining, non-orthogonal blades in LAP- and SAP-EPI (152,153). The use of reference scans (93) in the proposed phase-correction method typically increases the EPI-PROPELLER acquisition time by 2~3 TRs only, providing a significant improvement in time-efficiency over the use of blade-specific reference scan acquisitions, which can double the scan time of EPI-PROPELLER. The proposed correction also has the advantage of being able to reduce oblique phase errors using the linear phase errors obtained from standard, non-phase-encoded reference scans.

7.3 Future Work

The artifact reduction techniques described in Chapters 5 and 6 have demonstrated a significant reduction of the cusp artifact and the Nyquist ghost artifact, respectively. This section discusses possible areas of future development for both artifact reduction techniques.

In the implementation of *Aim 1*, Figure 5.24 showed limited improvements using the slice-tilting approach, when the gradient null (or, source of the artifact) falls in a region of large tissue heterogeneity. As mentioned in Section 5.8, motion compensation techniques may be applied to reduce the artifact under these conditions.

The slice tilting technique was applied towards the reduction of aliasing artifacts in PROPELLER imaging, using the same tilting angle for all blades of PROPELLER. The limitation of using a constant tilt angle in the current implementation is that the signal intensity of all PROPELLER blades is affected irrespective of their relative contributions to the artifacts, which is most likely responsible for the “center brightening” effect observed in Figure 5.22e-f. This problem may be solved by using a variable tilt angle scheme in which the optimal tilt angle is determined based on the blade orientation and the image geometry. It is also possible that the oblique tilt approach may play a larger role in these applications.

Aim 2 focused on the development and validation of the EPI-PROPELLER phase correction technique. However, PROPELLER based on EPI requires several optimization steps. Adjustments in data acquisition include partial k -space acquisitions and the use of parallel imaging to decrease the total imaging time. EPI-PROPELLER reconstruction therefore consists of additional steps such as homodyne processing (102) and SENSE (16) reconstruction, which are applied after phase correction. It will be important to ascertain that the phase correction step, which will be performed before these steps, does not introduce any phase inconsistencies in the data that interfere with reconstruction and post-processing.

Although the phase-correction technique has been discussed with reference to standard EPI-PROPELLER, it is relevant to any EPI-based acquisition sequence where the readout and phase-encoding gradients are combinations of the physical x , y , and z gradient axes. A direct extension of this work is in turboprop (122), which is a PROPELLER pulse sequence based on a gradient-and spin-echo (116,117) type acquisition strategy. This technique may also be applicable in 3D SAP-EPI (154,155), potentially allowing its extension to a “true” 3D acquisition, such as in the 3D overlapping “rod” acquisition method (156). Another possible extension of the phase-correction technique is in multi-slice cardiac or spine imaging, where each slice is aligned in a different plane, along or perpendicular to desired anatomical markers.

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Peer-Reviewed Journal Publication

1. **Novena Rangwala** and X. Joe Zhou, "Reduction of Fast Spin Echo Cusp Artifact Using A Slice-Tilting Gradient", *Magnetic Resonance in Medicine*, Jul. 2010, 64 (1): 220-228.

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Patent Applications

1. X. Joe Zhou and **Novena Rangwala**, “Method for Reducing Artifacts in Magnetic Resonance Imaging”, US provisional patent number 61/477,806, filed 21st April 2011.
2. X. Joe Zhou and **Novena Rangwala**, “Method for Reducing Artifacts in Magnetic Resonance Imaging”, international patent application number WO 2010/120829, filed 21st Oct. 2010, *pending*.

Selected Conference Publications

1. **Novena Rangwala** and X. Joe Zhou, “A Generalized Phase-Correction Technique for EPI-PROPELLER”, *Proc. 19th Ann. Meet. of the Intl. Soc. Magn. Reson. Med.*, Montreal, Canada, 2011, p.4613.
2. Girish Srinivasan, **Novena Rangwala**, X. Joe Zhou, “Reduction of Image Distortion in Non-Axial Diffusion-Weighted Imaging using Steer-PROP”, *Proc. 19th Ann. Meet. of the Intl. Soc. Magn. Reson. Med.*, Montreal, Canada, 2011, p.1955.
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4. Anoop Ganda, David Nyenhuis, Mira Rizkalla, Fu-qiang Gao, Jennifer Marola, Christopher Scott, Kie Honjo, **Novena Rangwala**, Laura Pedelty, Nancy Lobaugh, Simon Graham, Erin Gibson, X. Joe Zhou, Glenn Stebbins, Donald Stuss, Sandra Black, “Sensitivity and Specificity of Four Neuropsychological Screening Measures in an Ischemic Stroke Sample”, *accepted to the Canadian Stroke Congress*, 2011.
5. Christopher Scott, Kie Honjo, David Nyenhuis, Fu-qiang Gao, X. Joe Zhou, Simon Graham, **Novena Rangwala**, Glenn Stebbins, Erin Gibson, Anoop Ganda, Nancy Lobaugh, Jennifer Marola, Donald Stuss, Sandra Black, “Thalamic Volume is Related to Side and Location of Stroke, Stroke Volume, and Post-Stroke Cognitive Impairment”, *accepted to the Canadian Stroke Congress*, 2011.
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10. **Novena Rangwala** and X. Joe Zhou, "Reduction of FSE Cusp Artifact Using Slice Tilting", *Proc. 17th Ann. Meet. of the Intl. Soc. Magn. Reson. Med.*, Honolulu, HI, USA, 2009, p.2802.
11. **Novena Rangwala**, Girish Srinivasan, X. Joe Zhou, "Reduction of Aliasing Artifacts in Diffusion-Weighted PROPELLER Imaging", *Proc. 17th Ann. Meet. of the Intl. Soc. Magn. Reson. Med.*, Honolulu, HI, USA, 2009, p.2683.
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Awards and Activities

- ISMRM Member, 2003 – present.
- ISMRM Educational Stipend Award, Feb. 2009 – 2011.
- New Entrant Stipend Award from ISMRM, Mar. 2003.
- J. N. Tata Endowment for Graduate Studies, Jul. 2002.
- National Talent Search Scholarship for College Education, Jun. 1996.