EXAMPLE 23. Magnetic Resonance Imaging $\frac{1}{2}$ Wolfgang R. Nitz
Wolfgang R. Nitz **23. Magnetic Re**

Magnetic resonance imaging (MRI) represents

an exciting technology not only from a tech-

represents

an exciting technology not only from a tech-

23.3.1 Slice Select

23.3.2 The Spin-E

23.3.3 The Mul **23. Magnetic Reso**
 Polfgang R. Nitz

Magnetic resonance imaging (MRI) represents

an exciting technology not only from a tech-

nological perspective but also in view of it's

clinical potential. A brief introduction w **Example 16 23.3 MRI – Basic Principles and Applications ... 446

23.3 MRI – Basic Principles and Applications ... 446

23.3.1 Slice Selection and Spatial Encoding 446

23.3.2 The Spin-Echo Sequence** 23.3.1 Slice Selection and Spatial Encoding ⁴⁴⁶ 23.3.2 The Spin-Echo Sequence ⁴⁴⁷

**Example 18 Controlling Properties Are controlling Properties Are controlling Properties and exciting technology not only from a technological perspective but also in view of it's

given with respect to the historical deve** Volfgang R. Nitz

Magnetic resonance imaging (MRI) represents

an exciting technology not only from a tech-

nological perspective but also in view of it's

clinical potential. A brief introduction will be

given with resp **Example 16 The Historical develop-**

Magnetic resonance imaging (MRI) represents

an exciting technology not only from a tech-

nological perspective but also in view of it's

clinical potential. A brief introduction will Volfgang R. Nitz

Magnetic resonance imaging (MRI) represents

an exciting technology not only from a tech-

23.3.1 Slic

nological perspective but also in view of it's

clinical potential. A brief introduction will be

gi Verteingland R. Nitz

Magnetic resonance imaging (MRI) represents

an exciting technology not only from a tech-

mological perspective but also in view of it's

clinical potential. A brief introduction will be

given with Valiang R. Nitz

Magnetic resonance imaging (MRI) represents

an exciting technology not only from a tech-

nological perspective but also in view of it's

clinical potential. A brief introduction will be

given with respe Magnetic resonance imaging (MRI) represents

an exciting technology not only from a technological perspective but also in view of it's

clinical potential. A brief introduction will be

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an exciting technology not only from a tech-

nological perspective but also in view of it's

clinical potential. A brief introduction will be

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an exciting technology not only from a tech-

nological perspective but also in view of it's

clinical potential. A brief introduction will be

given with respect to the histor magnetic resonance imaging (MRI) represents

an exciting technology not only from a tech-

nological perspective but also in view of it's

given with respect to the historical develop-

ment of *nuclear magnetic resonance* an exciting technology not only from a tech-
nological perspective but also in view of it's
clinical potential. A brief introduction will be
given with respect to the historical develop-
ment of *nuclear magnetic resonance* mological perspective but also in view of it's

clinical potential. A brief introduction will be

given with respect to the historical develop-

ment of *nuclear magnetic resonance* (NMR)

23.3.3 The Gradient

later to be considered ative outlook to the future of MRI and combined
attack of the future considered primarily composed of a magnetic aspect as negative function of *nuclear magnetic resonance* (NMR)
and a magnetic field gradient sy given with respect to the historical develop-

ment of *nuclear magnetic resonance* (NMR)

later to be called MRI. A MRI system can be

considered primarily composed of a magnet,

a magnetic field gradient system, *radio f* ment of *nuclear magnetic resonance* (NMR)

later to be called MRI. A MRI system can be

23.3.5 The Sequence

considered primarily composed of a magnet,

a magnetic field gradient system, *radio frequency*

23.4.4 **MRI – S** modalities. (KF) colls for signal processing. The introduc-

tion of those hardware components is followed

by a description of basic MRI principles and

applications. Although MRI is a noninvasive

technology working without ionizing 23.2 MRI – System Components..................... ⁴⁴¹

for the Magnetic Field Gradient 442

23.2.3 The Radiofrequency System 444

23.5.5 Theranostics –

23.2.4 Measurement Control, Acquisition,

and Image Reconstruction Systems. 445 References gene Uhlenbeck and Samuel A. Goudsmit. In 1933, Otto Stern and Walther Gerlach were successful in **23.1 History of MRI**
23.1 History of MRI
23.1 History of MRI
23.1 History of MRI
23.1 History of MRI
in 1924 [23.1]. His statement followed a year after the Rabi of Columbia University in New
introduction of the **23.1 History of MRI**
Wolfgang Pauli proposed the existence of a nuclear spin field [23.2–5]. Supported by
in 1924 [23.1]. His statement followed a year after the Rabi of Columbia University
introduction of the spin of th

ful in measuring the nuclear magnetic moment [23.6] **Example 25.5.5 Interary Under Image Guidance 458**
 References
 Reference of Gorter,
 ments, but with negative result. Gorter was the first to use the term nuclear magnetic resonance (NMR) in his

It is general perception that the experiments launch-
the development of NMR were performed by *Bloch* tions [23.10] to characterize t
Purcell in 1946, proving the existence of the nu-
mechanism of recovery is called ti **Medical Imaging**
It is general perception that the experiments launch-
ing the development of NMR were performed by *Bloch* tions [23.10] to characterize
and *Purcell* in 1946, proving the existence of the nu-
mechanism **Medical Imaging**

It is general perception that the experiments launch-

ing the development of NMR were performed by *Bloch* tions [23.10] to characterize this

and *Purcell* in 1946, proving the existence of the nu-

c **Medical Imaging**

It is general perception that the experiments launch-

ing the development of NMR were performed by *Bloch*

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and *Purcell* in 1946, proving the existence of the nu-

mechan **Medical Imaging**

It is general perception that the experiments launch-

ing the development of NMR were performed by *Bloch* tions [23.10] to characterize this

and *Purcell* in 1946, proving the existence of the nu-

c **Medical Imaging**

It is general perception that the experiments launch-

the term T_1 with

ing the development of NMR were performed by *Bloch*

tions [23.10] to

and *Purcell* in 1946, proving the existence of the nu **It is general perception that the experiments launch-** the term T_1 within his phenon the development of NMR were performed by *Bloch* tions [23.10] to characterize *Purcell* in 1946, proving the existence of the numec

It is general perception that the experiments launch-

in gthe development of NMR were performed by *Bloch* tions [23.10] to characterize t

and *Purcell* in 1946, proving the existence of the numericanism of recovery is and *Purcell* in 1946, proving the existence of the nu-

clear spin and the phenomenon of (nuclear) magnetic correlated time constant is ter

resonance [(N)MR] [23.8,9], work for which Bloch and time. The precessing trans clear spin and the phenomenon of (nuclear) magnetic correlated the
resonance [(N)MR] [23.8,9], work for which Bloch and time. The p
Purcell shared the Nobel Prize in 1952. will induce
The spin is characterized by an angul The spin is characterized by an angular momen-

to the object. The stum combined with a magnetic moment. The spin is a tissue-specific time, which Bla

a quantum-mechanical entity. As a consequence, the derlying mechanism

magnetic moment of the proton (with spin quantum constant the T_2 relaxation number 1/2), exposed to an external magnetic field B_0 , tissue-specific relaxation of the direction of the field. Those positions are *Damad* number 1/2), exposed to an external magnetic field B_0 , tissue-specific relaxation tir
can only have two possible positions: parallel or antipar-
allel to the direction of the field. Those positions are $Damadian$ at the Dow can only have two possible positions: parallel or antipar-
allel to the direction of the field. Those positions are *Damadian* at the Downstate Medica
characterized by a difference in energy, with the parallel lyn and *H* allel to the direction of the field. Those positions are *Damadian* at the Downstate characterized by a difference in energy, with the parallel lyn and *Hollis* at Johns Hopki evaluated the T_1 and T_2 re If the energ characterized by a difference in energy, with the parallel lyn and *H*
position being energetically preferred. The evaluated is the energy of an electromagnetic wave corre-
sponds exactly to the energy difference between two possible states, parallel aligned spins can be pushed

into the antiparallel position. With the return of the an-

into the antiparallel position. With the return of the an-

the had found the ultimate techno

tiparal into the antiparallel position. With the return of the an-

tiparallel spins back to the original parallel position, cancer [23.14]. Unfortunately, the energy difference is released as an electromagnetic regarding how to

$$
\Delta E = \gamma \hbar B_0 = h v = h \frac{c}{\lambda} ,
$$

 $\Delta E = \gamma \hbar B_0 = h\nu = h \frac{c}{\lambda}$, posed on a static magnetic field the signal induced by the precessintization [23.15]. The previously metromenon is also called *magnetic resonance*. As more spins are aligned parallel to the $\Delta E = \gamma \hbar B_0 = h\nu = h\frac{1}{\lambda}$, the signal induced by the precessintization [23.15]. The previously metric in the spin induced by the precessintization [23.15]. The previously metric magnetic resonance condition, the phe-
f which is typical for a *resonance* condition, the phecional fraction (23.15). The previously m
nomenon is also called *magnetic resonance*. field. If a magnetic field gradient
As more spins are aligned parallel to the ext which is typical for a *resonance* condition, the phe-

frequency is a function of the strem

nomenon is also called *magnetic resonance*.

field. If a magnetic field gradient

As more spins are aligned parallel to the ex nomenon is also called *magnetic resonance*. field. If a magnetic field gradient
As more spins are aligned parallel to the external the static magnetic field, the magnet
magnetic field, as compared with the antiparallel As more spins are aligned parallel to the external
magnetic field, the magnetic field, as compared with the antiparallel align-
ment, a longitudinal nuclear magnetization builds up.
resonance frequency becomes a
The longi magnetic field, as compared with the antiparallel align-
ment, a longitudinal nuclear magnetization builds up.
mesonance frequency becomes a
The longitudinal nuclear magnetization can be treated Fourier transformation (FT ment, a longitudinal nuclear magnetization builds up.
The longitudinal nuclear magnetization can be treated Fourier transformation (1
using the principles of classical physics. If the longi-
the signal induced by th
tudin The longitudinal nuclear magnetization can be treated Fourier transformation (FT) calusing the principles of classical physics. If the longi-
the signal induced by the transverse included in the transverse nuclear magneti using the principles of classical physics. If the longi-
the signal induced by the trate
tudinal nuclear magnetization is tilted away from the
parallel direction of the external magnetic field, the
indicating the location tudinal nuclear magnetization is tilted away from the zation in order to ident
parallel direction of the external magnetic field, the indicating the location, a
angular momentum of the spins causes precession of to the pi parallel direction of the external magnetic field, the indicating the location, and thereby
angular momentum of the spins causes precession of to the pixel on the display represe
the magnetization around the original alig the magnetization around the original alignment (the proportional to the detected amposite difference of the external magnetic field B_0). This pre-
cessional frequency is called the Larmor frequency and demonstrate spa direction of the external magnetic field B_0). This precessional frequency is called the Larmor frequency and demonstrate spa
is identical to the frequency representing the energy onance (NMR)
difference between the two

Medical Imaging

It is general perception that the experiments launch-

in general perception that the experiments launch-

in the development of NMR were performed by *Bloch* tions [23.10] to characterize this *re*

an **Medical Imaging**

It is general perception that the experiments launch-

ing the development of NMR were performed by *Bloch* tions [23.10] to characterize this is

and *Purcell* in 1946, proving the existence of the nu-It is general perception that the experiments launch-
ing the development of NMR were performed by *Bloch* tions [23.10] to characterize
and *Purcell* in 1946, proving the existence of the nu-
nechanism of recovery is cal ing the development of NMR were performed by *Bloch* tions [23.10] to characterize this *i* and *Purcell* in 1946, proving the existence of the nu-
clear spin and the phenomenon of (nuclear) magnetic correlated time const mance [(N)MR] [23.8,9], work for which Bloch and

time. The precessing transverse icell shared the Nobel Prize in 1952.

Will induce a voltage in any ante

The spin is characterized by an angular momen-

close vicinity to Purcell shared the Nobel Prize in 1952. will induce a voltage in any anter

The spin is characterized by an angular momen-

close vicinity to the object. The sig

tum combined with a magnetic moment. The spin is a tissuetum combined with a magnetic moment. The spin is a tissue-specific time, which Bloch
a quantum-mechanical entity. As a consequence, the derlying mechanism is called T_2 rela
magnetic moment of the proton (with spin quan a quantum-mechanical entity. As a consequence, the derlying mechanism is called 7 magnetic moment of the proton (with spin quantum constant the T_2 relaxation time number 1/2), exposed to an external magnetic field B_0 the term T_1 within his phenomenological Bloch equations [23.10] to characterize this *recovery time*. The mechanism of recovery is called *relaxation*, and the correlated time constant is termed the T_2 , relaxation the term T_1 within his phenomenological Bloch equations [23.10] to characterize this *recovery time*. The mechanism of recovery is called *relaxation*, and the correlated time constant is termed the T_1 relaxation ti the term T_1 within his phenomenological Bloch equations [23.10] to characterize this *recovery time*. The mechanism of recovery is called *relaxation*, and the correlated time constant is termed the T_1 relaxation wi the term T_1 within his phenomenological Bloch equations [23.10] to characterize this *recovery time*. The mechanism of recovery is called *relaxation*, and the correlated time constant is termed the T_1 relaxation ti the term T_1 within his phenomenological Bloch equations [23.10] to characterize this *recovery time*. The mechanism of recovery is called *relaxation*, and the correlated time constant is termed the T_1 relaxation ti the term T_1 within his phenomenological Bloch equations [23.10] to characterize this *recovery time*. The mechanism of recovery is called *relaxation*, and the correlated time constant is termed the T_1 relaxation ti the term T_1 within his phenomenological Bloch equations [23.10] to characterize this *recovery time*. The mechanism of recovery is called *relaxation*, and the correlated time constant is termed the T_1 relaxation ti the term T_1 within his phenomenological Bloch equations [23.10] to characterize this *recovery time*. The mechanism of recovery is called *relaxation*, and the correlated time constant is termed the T_1 relaxation ti the term T_1 within his phenomenological Bloch equations [23.10] to characterize this *recovery time*. The mechanism of recovery is called *relaxation*, and the correlated time constant is termed the T_1 relaxation ti the term T_1 within his phenomenological Bloch equations [23.10] to characterize this *recovery time*. The mechanism of recovery is called *relaxation*, and the correlated time constant is termed the T_1 relaxation ti the term T_1 within his phenomenological Bloch equations [23.10] to characterize this *recovery time*. The mechanism of recovery is called *relaxation*, and the correlated time constant is termed the T_1 relaxation ti the term T_1 within his phenomenological Bloch equations [23.10] to characterize this *recovery time*. The mechanism of recovery is called *relaxation*, and the correlated time constant is termed the T_1 relaxation ti tions [23.10] to characterize this *recovery time*. The
mechanism of recovery is called *relaxation*, and the
correlated time constant is termed the T_1 relaxation
time. The precessing transverse nuclear magnetization
w mechanism of recovery is called *relaxation*, and the correlated time constant is termed the T_1 relaxation time. The precessing transverse nuclear magnetization will induce a voltage in any antenna system or coil in cl correlated time constant is termed the T_1 relaxation
time. The precessing transverse nuclear magnetization
will induce a voltage in any antenna system or coil in
close vicinity to the object. The signal will decay with time. The precessing transverse nuclear magnetization
will induce a voltage in any antenna system or coil in
close vicinity to the object. The signal will decay within
a tissue-specific time, which Bloch termed T_2 . The will induce a voltage in any antenna system or coil in
close vicinity to the object. The signal will decay within
a tissue-specific time, which Bloch termed T_2 . The un-
derlying mechanism is called T_2 relaxation, an close vicinity to the object. The signal will decay within
a tissue-specific time, which Bloch termed T_2 . The un-
derlying mechanism is called T_2 relaxation, and the time
constant the T_2 relaxation times on livin a tissue-specific time, which Bloch termed T_2 . The un-
derlying mechanism is called T_2 relaxation, and the time
constant the T_2 relaxation time. Experiments to measure
tissue-specific relaxation times on living c derlying mechanism is called T_2 relaxation, and the time
constant the T_2 relaxation time. Experiments to measure
tissue-specific relaxation times on living cells and ani-
mal tissues were performed as early as 1955 constant the T_2 relaxation time. Experiments to measure
tissue-specific relaxation times on living cells and ani-
mal tissues were performed as early as 1955 [23.11].
Damadian at the Downstate Medical Center in Brook-
 body. tissues were performed as early as 1955 [23.11].
 nadian at the Downstate Medical Center in Brook-

and *Hollis* at Johns Hopkins University in Baltimore

luated the T_1 and T_2 relaxation times of normal

cancer ti Damadian at the Downstate Medical Center in Brook-
lyn and *Hollis* at Johns Hopkins University in Baltimore
evaluated the T_1 and T_2 relaxation times of normal
and cancer tissue and observed that cancerous tissue
de lyn and *Hollis* at Johns Hopkins University in Baltimore
evaluated the T_1 and T_2 relaxation times of normal
and cancer tissue and observed that cancerous tissue
demonstrated longer relaxation times as compared with evaluated the T_1 and T_2 relaxation times of normal
and cancer tissue and observed that cancerous tissue
demonstrated longer relaxation times as compared with
normal tissue [23.12,13]. It was *Damadian*'s belief that entific metallisation in the entropy of the set of the s

the energy difference is released as an electromagnetic regarding how to perform cancer
wave, also called the NMR signal. As the wavelength body.
 λ of the stimulating electromagnetic field has to be
 λ the antipara wave, also called the NMR signal. As the wavelength body.
 λ of the stimulating electromagnetic field has to be

identical to the wavelength representing the difference

University of New York who, in 19

between the t λ of the stimulating electromagnetic field has to be

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University of New York who, in

between the two possible states
 $\Delta E = \gamma h B_0 = h v = h \frac{c}{\lambda}$,

tization [23.1 identical to the wavelength representing the difference University of New York who, in
between the two possible states concept of using a magnetic fiel
posed on a static magnetic fiel
the signal induced by the prece-
whic between the two possible states
 $\Delta E = \gamma h B_0 = h v = h \frac{c}{\lambda}$,

which is typical for a *resonance* condition, the phe-

frequency is direction in the phe-

numenon is also called *magnetic resonance*.

As more spins are alig and cancer tissue and observed that cancerous tissue
demonstrated longer relaxation times as compared with
normal tissue [23.12,13]. It was *Damadian*'s belief that
he had found the ultimate technology for diagnosing
canc demonstrated longer relaxation times as compared with
normal tissue [23.12, 13]. It was *Damadian*'s belief that
he had found the ultimate technology for diagnosing
cancer [23.14]. Unfortunately, he gave no practical hint
 normal tissue [23.12, 13]. It was *Damadian*'s belief that
he had found the ultimate technology for diagnosing
cancer [23.14]. Unfortunately, he gave no practical hint
regarding how to perform cancer screening on a human
b the had found the ultimate technology for diagnosing
cancer [23.14]. Unfortunately, he gave no practical hint
regarding how to perform cancer screening on a human
body.
It was the American chemist *Lauterbur* at the State
 cancer [23.14]. Unfortunately, he gave no practical hint
regarding how to perform cancer screening on a human
body.
It was the American chemist *Lauterbur* at the State
University of New York who, in 1971, introduced the
 regarding how to perform cancer screening on a human
body.
It was the American chemist *Lauterbur* at the State
University of New York who, in 1971, introduced the
concept of using a magnetic field gradient superim-
posed body.
It was the American chemist *Lauterbur* at the State
University of New York who, in 1971, introduced the
concept of using a magnetic field gradient superim-
posed on a static magnetic field to spatially encode
the si It was the American chemist *Lauterbur* at the State
University of New York who, in 1971, introduced the
concept of using a magnetic field gradient superim-
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the signal University of New York who, in 1971, introduced the
concept of using a magnetic field gradient superim-
posed on a static magnetic field to spatially encode
the signal induced by the precessing nuclear magne-
tization [23. concept of using a magnetic field gradient superim-
posed on a static magnetic field to spatially encode
the signal induced by the precessing nuclear magne-
tization [23.15]. The previously mentioned resonance
frequency is posed on a static magnetic field to spatially encode
the signal induced by the precessing nuclear magne-
tization [23.15]. The previously mentioned resonance
frequency is a function of the strength of the magnetic
field. I the signal induced by the precessing nuclear magne-
tization [23.15]. The previously mentioned resonance
frequency is a function of the strength of the magnetic
field. If a magnetic field gradient is superimposed on
the st tization [23.15]. The previously mentioned resonance
frequency is a function of the strength of the magnetic
field. If a magnetic field gradient is superimposed on
the static magnetic field, the magnetic field strength bequency is a function of the strength of the magnetic
d. If a magnetic field gradient is superimposed on
static magnetic field, the magnetic field strength be-
hes a function of location, and with this also the
phance frequ field. If a magnetic field gradient is superimposed on
the static magnetic field, the magnetic field strength be-
comes a function of location, and with this also the
resonance frequency becomes a function of location.
Fou the static magnetic field, the magnetic field strength be-
comes a function of location, and with this also the
resonance frequency becomes a function of location.
Fourier transformation (FT) can be applied to analyze
the comes a function of location, and with this also the
resonance frequency becomes a function of location.
Fourier transformation (FT) can be applied to analyze
the signal induced by the transverse nuclear magneti-
zation in

resonance frequency becomes a function of location.
Fourier transformation (FT) can be applied to analyze
the signal induced by the transverse nuclear magneti-
zation in order to identify the frequency components
indicatin Fourier transformation (FT) can be applied to analyze
the signal induced by the transverse nuclear magneti-
zation in order to identify the frequency components
indicating the location, and thereby assign a brightness
to t the signal induced by the transverse nuclear magneti-
zation in order to identify the frequency components
indicating the location, and thereby assign a brightness
to the pixel on the display representing that location,
pr zation in order to identify the frequency components
indicating the location, and thereby assign a brightness
to the pixel on the display representing that location,
proportional to the detected amplitude.
Lauterbur used t indicating the location, and thereby assign a brightness
to the pixel on the display representing that location,
proportional to the detected amplitude.
Lauterbur used two test-tubes filled with water to
demonstrate spatia tubes. portional to the detected amplitude.

Lauterbur used two test-tubes filled with water to

nonstrate spatially encoded nuclear magnetic res-

nce (NMR) by superimposing a magnetic field

lient on a static magnetic field, pr

Magnetic Resonance Imaging 23.2
also attended by Richard Ernst from the University of first x-ray computer tomography sy
Zurich. Ernst recognized that the back-projection could in 1976 the development of a scanne
be replac Magnetic Resonance Imaging

also attended by Richard Ernst from the University of first x-ray computer tomograph:

Zurich. Ernst recognized that the back-projection could in 1976 the development of a sc

be replaced by a c Magnetic Resonance Imagin

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Zurich. Ernst recognized that the back-projection could in 1976 the development of

be replaced by a combinat Magnetic Resonance Imaging 23
also attended by Richard Ernst from the University of first x-ray computer tomography
Zurich. Ernst recognized that the back-projection could in 1976 the development of a scar
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Zurich. Ernst recognized that the back-projection could in 1976 the development of a scann
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also attended by Richard Ernst from the University of first x-ray computer to
Zurich. Ernst recognized that the back-projection could in 1976 the developme
be replaced by a combination of phase and fre Magnetic Resonance Imaging

also attended by Richard Ernst from the University of first x-ray computer tomogra

Zurich. Ernst recognized that the back-projection could in 1976 the development of a

be replaced by a combina also attended by Richard Ernst from the University of
Zurich. Ernst recognized that the back-projection could
be replaced by a combination of phase and frequency
encoding of the MR signal [23.16]. This method is still
the Magnetic Resonance Im

Magnetic Resonance Im

in attended by Richard Ernst from the University of first x-ray computer to

ich. Ernst recognized that the back-projection could in 1976 the developmen

replaced by a combinat also attended by Richard Ernst from the University of first x-ray computer tomography s

Zurich. Ernst recognized that the back-projection could in 1976 the development of a scan

be replaced by a combination of phase and Zurich. Ernst recognized that the back-projection could in 1976 the to
be replaced by a combination of phase and frequency waves, and in
encoding of the MR signal [23.16]. This method is still head using M
the main recons encoding of the MR signal [23.16]. This method is still head using MRI. This annot
the main reconstruction algorithm used in NMR. To boost to the medical imaging
avoid the fear associated with the word *nuclear*, NMR to r

Strength B_0

Section 1981.

See the British company EMI, which revolutionized installations based on commercial procedure maging in 1973 with the introduction of the for routine clinical applications were medical imaging in 1973 with The British company EMI, which revolutionized installations based on commercial pro-
 STARI – System Components
 STARI – System Components
 STARI – System Components
 STARI – System Components
 STARI – System Com medical imaging in 1973 with the introduction of the for routine clinical applications
 23.2.1 The Magnet – The Magnetic Field The transverse nuclear magnetic **Strength** B_0 erated by tilting the longitudin

The pheno **23.2 MRI – System Components**
 23.2.1 The Magnet – The Magnetic Field The transverse nuclear magnetic **Strength** B_0 erated by ultimg the longitudina
 M_z away from the direction of t

served in the presence of a **23.2.1 The Magnet – The Magnetic Field**
 23.2.1 The Magnet – The Magnetic Field
 Strength B_0

The phenomenon of magnetic resonance is only ob-

served in the presence of a strong magnetic field. The

strength of t **Stem Components**
 and The Magnetic Field The transverse nuclear magnetizz
 B₀ erated by tilting the longitudinal nucle
 a f magnetic resonance is only ob- If a 90° RF excitation pulse is used, a

ce of a strong m **23.2 MRI – System Components**
 23.2.1 The Magnet – The Magnetic Field The transverse nu
 Strength B_0 erated by tilting the lot

The phenomenon of magnetic resonance is only ob- If a 90° RF excitation

served in th **2.1 The Magnet – The Magnetic Field** The transverse nuclear magnetical **Strength** B_0 The transverse nuclear magnetic magnetic potential M_z away from the direction of the phenomenon of magnetic resonance is only ob- **23.2.1 The Magnet – The Magnetic Field** The transverse nuclear is only contracted by tilting the longitude of a strong magnetic resonance is only ob- If a 90° RF excitation pulse served in the presence of a strong magnet **Strength B₀** erated by tilting the longitudinal
 M_z away from the direction of th

red in the presence of a strong magnetic field. The longitudinal nuclear magnetizatio

ed in the presence of a strong magnetic fiel The phenomenon of magnetic resonance is only ob-
served in the presence of a strong magnetic field. The longitudinal nuclear magnetizatist
strength of the static magnetic field B_0 is described in transverse nuclear mag The phenomenon of magnetic resonance is only ob-
served in the presence of a strong magnetic field. The longitudinal nuclear magnetization
strength of the static magnetic field B_0 is described in transverse nuclear mag

served in the presence of a strong magnetic field. The longitudinal nuclear magnetization
strength of the static magnetic field B_0 is described in transverse nuclear magnetization Λ
units of Tesla (T), at least with strength of the static magnetic field B_0 is described in transverse nuclear magnetization Λ
units of Tesla (T), at least within the medical commu-
in units of Vs/m² and the magnetic flux density trization is a fun units of Tesla (T), at least within the medical commu-
inty [for physicists, T means the magnetic flux density
in units of V s/m² and the magnetic field strength is the different energy levels (of paral
measured in unit mity [for physicists, *T* means the magnetic flux density tization is a function of the occurred in units of N/m or oerstedt (Oe)].

In magnetic field strength is the different energy levels (of measured in units of A/m o share. sured in units of A/m or oerstedt (Oe)]. lel alignment of the nuclear s
Early systems (1983) used resistive magnets with magnetic moments). The amp
metic field strength in the range 0.1–0.2 T. where the magnetization M_z Early systems (1983) used resistive magnets with magnetic moments). The amplitude
magnetic field strength in the range 0.1–0.2 T. muclear magnetization M_z scales wit
conductivity could be used to generate higher magneti magnetic field strength in the range 0.1–0.2 T.

Within the same year it became obvious that super-

strength used

conductivity could be used to generate higher magnetic

fields. Since then, the field strength used in cl Within the same year it became obvious that super-

strength used

conductivity could be used to generate higher magnetic

fields. Since then, the field strength used in clini-

cal scanners has steadily increased. In 198 conductivity could be used to generate higher magneticals. Since then, the field strength used in c
cal scanners has steadily increased. In 1985, the
strength of the majority of the systems was 0.5 T.
rently, in 2010, t 15. Since then, the field strength used in clini-

scanners has steadily increased. In 1985, the field Considering amplitude and frequen

ngth of the majority of the systems was 0.5 T. Cur-

nal is proportional to the squ cal scanners has steadily increased. In 1985, the field Considering amplitude and frequency
strength of the majority of the systems was 0.5 T. Cur-
nal is proportional to the square of
rently, in 2010, there is a strong t

strength of the majority of the systems was 0.5 T. Currently, in 2010, there is a strong tendency towards 3.0 T
systems, whereas 1.5 T systems have the largest market
share.
The primary reason for using a higher magnetic
 Share.

The primary reason for using a higher magnetic and the signal-to-noise ratio (field strength is obvious: the ratio between signal and linear in the magnetic field strength a first approximation, proportional to th

$$
N\sim B_0
$$

$$
S \sim \frac{dM_{xy}}{dt} \sim vM_{xy},
$$

$$
v \sim \gamma B_0.
$$

Magnetic Resonance Imaging 23.2 MRI – System Components
first x-ray computer tomography system, announced
in 1976 the development of a scanner based on radio
waves, and in 1978 presented the first image of a human
head us Magnetic Resonance Imaging | 23.2 MRI – System Components

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head us first x-ray computer tomography system, announced
in 1976 the development of a scanner based on radio
waves, and in 1978 presented the first image of a human
head using MRI. This announcement gave a significant
boost to th d using MRI. This announcement gave a significant
st to the medical imaging industry, as they were not
epeat their mistake of underestimating the poten-
of a new imaging modality as they did with x-ray
putter tomography. boost to the medical imaging industry, as they were not
to repeat their mistake of underestimating the poten-
tial of a new imaging modality as they did with x-ray
computer tomography. In the early 1980s, the first MR
ins Negretic Resonance imaging

in 1974 (in the program) system. amounted that the product of a scanner hased on adio

in 1976 (in development of a scanner hased on adio

waves, and in 1978 (pseudent diving the product Part C

was more and more called magnetic resonance imaging tial of a new imaging modality as

(MRI) around 1981.

The British company EMI, which revolutionized installations based on commercial p

medical imaging in 1973 with th to repeat their mistake of underestimating the potential of a new imaging modality as they did with x-ray
computer tomography. In the early 1980s, the first MR
installations based on commercial products and aiming
for rou It also a new imaging modality as they did with x-ray

computer tomography. In the early 1980s, the first MR

installations based on commercial products and aiming

for routine clinical applications were seen.

The transv computer tomography. In the early 1980s, the first MR
installations based on commercial products and aiming
for routine clinical applications were seen.
The transverse nuclear magnetization M_x is gen-
erated by tilting is converted to installations based on commercial products and aiming
for routine clinical applications were seen.
The transverse nuclear magnetization M_x is gen-
erated by tilting the longitudinal nuclear magnetization
 M_z away from The transverse nuclear magnetization M_{xy} is generated by tilting the longitudinal nuclear magnetization M_z away from the direction of the main magnetic field.
If a 90° RF excitation pulse is used, all of the availabl The transverse nuclear magnetization M_{xy} is generated by tilting the longitudinal nuclear magnetization M_z away from the direction of the main magnetic field.
If a 90° RF excitation pulse is used, all of the availabl

The transverse nuclear magnetization M_{xy} is generated by tilting the longitudinal nuclear magnetization M_z away from the direction of the main magnetic field.
If a 90° RF excitation pulse is used, all of the availabl The transverse nuclear magnetization M_{xy} is generated by tilting the longitudinal nuclear magnetization M_z away from the direction of the main magnetic field. If a 90° RF excitation pulse is used, all of the availabl The transverse nuclear magnetization M_{xy} is
erated by tilting the longitudinal nuclear magnetiz
 M_z away from the direction of the main magnetic
If a 90° RF excitation pulse is used, all of the ava
longitudinal nuclea r magnetization M_{xy} is gen-
udinal nuclear magnetization
n of the main magnetic field.
se is used, all of the available
netization M_z is converted to
ization M_{xy} .
longitudinal nuclear magne-
he occupation probabil erated by tilting the longitudinal nuclear magnetiza M_z away from the direction of the main magnetic If a 90° RF excitation pulse is used, all of the avail longitudinal nuclear magnetization M_z is converted transverse longitudinal nuclear magnetization M_z is converted to
transverse nuclear magnetization M_{xy} .
The amplitude of the longitudinal nuclear magne-
tization is a function of the occupation probability of
the different energ transverse nuclear magnetization M_{xy} .
The amplitude of the longitudinal nuclear magne-
tization is a function of the occupation probability of
the different energy levels (of parallel and antiparal-
lel alignment of th The amplitude of the longitudinal nuclear matization is a function of the occupation probabilit
the different energy levels (of parallel and antip
lel alignment of the nuclear spins and their correl
magnetic moments). The for any and the matter spins and their correlated
magnetic moments). The amplitude of the longitudinal
nuclear magnetization M_z scales with the magnetic field
strength used
 $M_z \sim B_0$.
Considering amplitude and frequency

$$
M_z \sim B_0.
$$

magnetic moments). The amplitude of the longitudinal
nuclear magnetization M_z scales with the magnetic field
strength used
 $M_z \sim B_0$.
Considering amplitude and frequency, the induced sig-
nal is proportional to the squa Considering amplitude and frequency, the induced sig-
nal is proportional to the square of the magnetic field
strength used
 $S \sim B_0^2$,
and the signal-to-noise ratio (SNR) is approximately
linear in the magnetic field str

$$
S \sim B_0^2 \; ,
$$

$$
SNR = \frac{S}{N} \sim B_0 f(T_1, T_2)
$$

The primary reason for using a nigher magnetic and the signal-to-noise ratio (field strength is obvious: the ratio between signal and linear in the magnetic field strend has a first approximation, proportional to the magn The integral to the signal to-noise ratio between signal and linear in the magnetic fiel
background noise (the signal-to-noise ratio, SNR) is, to
distant approximation, proportional to the magnetic field
strength [23.17]. Considering amplitude and frequency, the induced sig-

nal is proportional to the square of the magnetic field

strength used
 $S \sim B_0^2$,

and the signal-to-noise ratio (SNR) is approximately

linear in the magnetic fiel strength used
 $S \sim B_0^2$,

and the signal-to-noise ratio (SNR) is approximately

linear in the magnetic field strength
 $SNR = \frac{S}{N} \sim B_0 f (T_1, T_2)$

and is of course a function of the tissue-specific relax-

ation times $S \sim B_0^2$,
and the signal-to-noise ratio (SNR) is approximately
linear in the magnetic field strength
SNR = $\frac{S}{N} \sim B_0 f(T_1, T_2)$
and is of course a function of the tissue-specific relax-
ation times T_1 and T_2 .
T $S \sim B_0^-$,
and the signal-to-noise ratio (SNR) is approximately
linear in the magnetic field strength
 $SNR = \frac{S}{N} \sim B_0 f(T_1, T_2)$
and is of course a function of the tissue-specific relax-
ation times T_1 and T_2 .
The i and the signal-to-noise ratio (SNR) is approximately
linear in the magnetic field strength
 $SNR = \frac{S}{N} \sim B_0 f(T_1, T_2)$
and is of course a function of the tissue-specific relax-
ation times T_1 and T_2 .
The increase of linear in the magnetic field strength
 $SNR = \frac{S}{N} \sim B_0 f (T_1, T_2)$

and is of course a function of the tissue-specific relax-

ation times T_1 and T_2 .

The increase of the SNR is only one aspect of
 $high-field$ systems. The SNR = $\frac{S}{N} \sim B_0 f(T_1, T_2)$
and is of course a function of the tissue-specific relax-
ation times T_1 and T_2 .
The increase of the SNR is only one aspect of
high-field systems. There are a number of phenomena
repr SNR = $\frac{3}{N} \sim B_0 f(T_1, T_2)$
and is of course a function of the tissue-specific relax-
ation times T_1 and T_2 .
The increase of the SNR is only one aspect of
high-field systems. There are a number of phenomena
repr *N*
and is of course a function of the tissue-specific relax-
ation times T_1 and T_2 .
The increase of the SNR is only one aspect of
high-field systems. There are a number of phenomena
representing advantages and di

Fig. 23.1a, b A commer-
cially available MR system
using a permanent mag-
net with a magnetic field Fig. 23.1a, b A commer-
cially available MR system
using a permanent mag-
net with a magnetic field
strength of 0.35 T (MAGNE-Fig. 23.1a, b A commer-
cially available MR system
using a permanent mag-
net with a magnetic field
strength of 0.35 T (MAGNE-
TOM C¹). The *T*₁ weighted Fig. 23.1a, b A commer-
cially available MR system
using a permanent mag-
net with a magnetic field
strength of 0.35 T (MAGNE-
TOM C!). The T_1 -weighted
sagittal image of a human Fig. 23.1a, b A commer-
cially available MR system
using a permanent mag-
net with a magnetic field
strength of 0.35 T (MAGNE-
TOM C!). The T_1 -weighted
sagittal image of a human
hrain demonstrates a slice Fig. 23.1a, b A commer-
cially available MR system
using a permanent mag-
net with a magnetic field
strength of 0.35 T (MAGNE-
TOM C!). The T_1 -weighted
sagittal image of a human
brain demonstrates a slice
thickness of 4 Fig. 23.1a, b A commer-
cially available MR system
using a permanent mag-
net with a magnetic field
strength of 0.35 T (MAGNE-
TOM C!). The T_1 -weighted
sagittal image of a human
brain demonstrates a slice
thickness of Fig. 23.1a, b A commer-
cially available MR system
using a permanent mag-
net with a magnetic field
strength of 0.35 T (MAGNE-
TOM C!). The T_1 -weighted
sagittal image of a human
brain demonstrates a slice
thickness o Fig. 23.1a, b A commer-
cially available MR system
using a permanent mag-
net with a magnetic field
strength of 0.35 T (MAGNE-
TOM C!). The T_1 -weighted
sagittal image of a human
brain demonstrates a slice
thickness o Fig. 23.1a, b A commer-
cially available MR system
using a permanent mag-
net with a magnetic field
strength of 0.35 T (MAGNE-
TOM C!). The T_1 -weighted
sagittal image of a human
brain demonstrates a slice
thickness of 4 Fig. 23.1a, b A commer-
cially available MR system
using a permanent mag-
net with a magnetic field
strength of 0.35 T (MAGNE-
TOM C!). The T_1 -weighted
sagittal image of a human
brain demonstrates a slice
thickness o Fig. 23.1a, b A commer-
cially available MR system
using a permanent mag-
net with a magnetic field
strength of 0.35 T (MAGNE-
TOM C!). The T_1 -weighted
sagittal image of a human
brain demonstrates a slice
thickness of

Beijing
of magnetic field strength and magnetic susceptibility. field strength that does not chan
With the improvement in SNR, motion artifacts are and that the generation of any reas
no longer masked by noise and are ofte of magnetic field strength and magnetic susceptibility. field strength that
With the improvement in SNR, motion artifacts are and that the genera
no longer masked by noise and are often pronounced strength seems to
due to magnetic field strength and magnetic susceptibility. field strength that does not channon the improvement in SNR, motion artifacts are and that the generation of any reason of the major masked by noise and are often pronou

due to the strong signal from fat-containing moving weight of the magnets is motissue. Safety-relevant aspects need special considera-
system). The advantages at
tion [23.18] as some potentially hazardous interactions ing tissue. Safety-relevant aspects need special considera-

system). The advantages and disadv

tion [23.18] as some potentially hazardous interactions

ing with higher magnetic field streng

for an MR system scale with fiel tion [23.18] as some potentially hazardous interactions ing with higher magnetic field scale with the field strength. Last but not least, the costs discussed in conjunction with sfor an MR system scale with field strength scale with the field strength. Last but not least, the costs
for an MR system scale with field strength.
Besides the magnetic field st
Around 1991 the major vendors decided to intro-
plays an important role that
duce lowfor an MR system scale with field strength.

Besides the magnetic field strength

Around 1991 the major vendors decided to intro-

plays an important role that should

duce low-cost *low-field* MR systems (field strength Around 1991 the major vendors decided to intro-
duce low-cost *low-field* MR systems (field strength up
mated. The size of the patient
to 0.4 T) parallel to the *high-field* systems, as 1.0 and
60 cm in the past, mainly fo duce low-cost *low-field* MR systems (field strength up mated. The size of the patient bore
to 0.4 T) parallel to the *high-field* systems, as 1.0 and 60 cm in the past, mainly for financia
1.5 T systems were called at th to 0.4 T) parallel to the *high-field* systems, as 1.0 and 60 cm in the past, mainly for fina
1.5 T systems were called at that time. This diversity has since 2004 systems with larger
since been maintained, and *low-field* 1.5 T systems were called at that time. This diversity has since 2004 systems with larger p
since been maintained, and *low-field* systems are still have been available (e.g., 70 cm; 1
commercially available (Fig. 23.1). since been maintained, and *low-field* systems are still have been available (e.g., 70 cm; commercially available (Fig. 23.1). A magnetic field opening of the patient bore is not strength of up to 0.35 T can be achieved b commercially available (Fig. 23.1). A magnetic field opening of the patient bore is not on
strength of up to 0.35 T can be achieved by assembling tor for the patient; it is also helpfor
a number of permanent magnets. Only strength of up to 0.35 T can be achieved by assembling tor for the patient; it is als
a number of permanent magnets. Only one vendor is number of medical examina
known to be still working with resistive magnets (stand-tro a number of permanent magnets. Only one vendor is

known to be still working with resistive magnets (stand-

trophobia and provides a poss

up MRI, FONAR Corp.). All other vendors (Siemens, whose circumference would n

Ge known to be still working with resistive magnets (stand-

up MRI, FONAR Corp.). All other vendors (Siemens, whose circumference would need

General Electric, Philips, Toshiba etc.) are working with bore.

either permanent up MRI, FONAR Corp.). All other vendors (Siemens, whose circumference would need n
General Electric, Philips, Toshiba etc.) are working with bore.

either permanent magnets or superconductive magnets.

Production and main General Electric, Philips, Toshiba etc.) are working with

either permanent magnets or superconductive magnets.

Production and maintenance of permanent systems are

relatively cost effective. A disadvantage is the limita either permanent magnets or superconductive magnets.

Production and maintenance of permanent systems are

relatively cost effective. A disadvantage is the limita-

for the Magnetic Fiel

tion to low field strength and th Production and maintenance of permanent systems are **23.2.2 The System**
relatively cost effective. A disadvantage is the limita-
tion to low field strength and the corresponding limit
in SNR. Another disadvantage is the c relatively cost effective. A disadvantage is the limita-
tion to low field strength and the corresponding limit
in SNR. Another disadvantage is the change in mag-
netic field strength as a function of temperature of the M tion to low field strength and the corresponding limit
in SNR. Another disadvantage is the change in mag-
netic field strength as a function of temperature of the MR resonance frequency, also ca
the magnet material. Appro in SNR. Another disadvantage is the change in mag-

in SNR. Another disadvantage is the change in mag-

To excite a specific slice or vol

the magnet material. Approximately 14 t of permanent quency, has to be a function **APTE** Notice in the set of the se

of magnetic field strength and magnetic susceptibility. field strength that does not change
With the improvement in SNR, motion artifacts are and that the generation of any reasons
no longer masked by noise and are often of magnetic field strength and magnetic susceptibility. field strength that does not chang
With the improvement in SNR, motion artifacts are and that the generation of any reason
no longer masked by noise and are often pr of magnetic field strength and magnetic susceptibility. field strength that does not change
With the improvement in SNR, motion artifacts are and that the generation of any reasor
no longer masked by noise and are often p With the improvement in SNR, motion artifacts are and that the generation of any reaso
no longer masked by noise and are often pronounced strength seems to be possible with
due to the strong signal from fat-containing mov no longer masked by noise and are often pronounced

the due to the strong signal from fat-containing moving

weight of the magnets is moderate

tissue. Safety-relevant aspects need special considerasystem). The advantag brain demonstrates a slice
thickness of 4 mm, acquired
within 5 min (courtesy Air-
force General Hospital,
Beijing, PR China)
force General Hospital,
Beijing, PR China)
seems to be possible without limit and that the
weigh thickness of 4 mm, acquired
within 5 min (courtesy Air-
force General Hospital,
Beijing, PR China)
Beijing, PR China)
force General Hospital,
Beijing, PR China)
about 1.5 T system and that the
weight of the magnets is mode within 5 min (courtesy Air-
force General Hospital,
Beijing, PR China)
field strength that does not change with temperature
and that the generation of any reasonable magnetic field
strength seems to be possible without lim force General Hospital,

Beijing, PR China)

field strength that does not change with temperature

and that the generation of any reasonable magnetic field

strength seems to be possible without limit and that the

weight Beijing, PR China)
field strength that does not change with temperature
and that the generation of any reasonable magnetic field
strength seems to be possible without limit and that the
weight of the magnets is moderate (a field strength that does not change with temperature
and that the generation of any reasonable magnetic field
strength seems to be possible without limit and that the
weight of the magnets is moderate (about 4 t for a 1.5 field strength that does not change with temperature
and that the generation of any reasonable magnetic field
strength seems to be possible without limit and that the
weight of the magnets is moderate (about 4 t for a 1.5 field strength that does not change with temperature
and that the generation of any reasonable magnetic field
strength seems to be possible without limit and that the
weight of the magnets is moderate (about 4 t for a $1.$ field strength that does not change with temperature
and that the generation of any reasonable magnetic field
strength seems to be possible without limit and that the
weight of the magnets is moderate (about 4 t for a 1.5 field strength that does not change with temperature
and that the generation of any reasonable magnetic field
strength seems to be possible without limit and that the
weight of the magnets is moderate (about 4 t for a 1.5 and that the generation of any reasonable magnetic field
strength seems to be possible without limit and that the
weight of the magnets is moderate (about 4 t for a 1.5 T
system). The advantages and disadvantages of work-
 strength seems to be possible without limit and that the
weight of the magnets is moderate (about 4 t for a 1.5 T
system). The advantages and disadvantages of work-
ing with higher magnetic field strength will be further
d weight of the magnets is moderate (about 4 t for a 1.5 T
system). The advantages and disadvantages of work-
ing with higher magnetic field strength will be further
discussed in conjunction with safety-relevant aspects.
Bes system). The advantages and disadvantages of work-
ing with higher magnetic field strength will be further
discussed in conjunction with safety-relevant aspects.
Besides the magnetic field strength, the magnet design
plays ing with higher magnetic field strength will be further discussed in conjunction with safety-relevant aspects.
Besides the magnetic field strength, the magnet design plays an important role that should not be underestimate discussed in conjunction with safety-relevant aspects.
Besides the magnetic field strength, the magnet design
plays an important role that should not be underesti-
mated. The size of the patient bore has been limited to
60 bore. mated. The size of the patient bore has been limited to 60 cm in the past, mainly for financial reasons, although since 2004 systems with larger patient bore openings have been available (e.g., 70 cm; Fig. 23.2). The larg the past, mainly for financial reasons, although
04 systems with larger patient bore openings
n available (e.g., 70 cm; Fig. 23.2). The larger
of the patient bore is not only a comfort fac-
he patient; it is also helpful i have been available (e.g., 70 cm; Fig. 23.2). The larger
opening of the patient bore is not only a comfort fac-
tor for the patient; it is also helpful in reducing the
number of medical examinations refused due to claus-
 opening of the patient bore is not only a comfort fac-
tor for the patient; it is also helpful in reducing the
number of medical examinations refused due to claus-
trophobia and provides a possibility to study patients
who

tor for the patient; it is also helpful in reducing the
number of medical examinations refused due to claus-
trophobia and provides a possibility to study patients
whose circumference would need more than a 60 cm
bore.
2 mumber of medical examinations refused due to claus-
trophobia and provides a possibility to study patients
whose circumference would need more than a 60 cm
bore.
23.2.2 The System
for the Magnetic Field Gradient
To exci trophobia and provides a possibility to study patients
whose circumference would need more than a 60 cm
bore.
23.2.2 The System
for the Magnetic Field Gradient
To excite a specific slice or volume at a specific locatio whose circumference would need more than a 60 cm
bore.
 23.2.2 The System
 for the Magnetic Field Gradient

To excite a specific slice or volume at a specific location,

the MR resonance frequency, also called the Larm **23.2.2 The System**
 for the Magnetic Field Gradient

To excite a specific slice or volume at a specific location,

the MR resonance frequency, also called the Larmor fre-

quency, has to be a function of location. For **23.2.2 The System**

for the Magnetic Field Gradient

To excite a specific slice or volume at a specific location,

the MR resonance frequency, also called the Larmor fre-

quency, has to be a function of location. For th **23.2.2 The System**
for the Magnetic Field Gradient
To excite a specific slice or volume at a specific location,
the MR resonance frequency, also called the Larmor fre-
quency, has to be a function of location. For this r **For the Magnetic Field Gradient**
To excite a specific slice or volume at a specific location,
the MR resonance frequency, also called the Larmor fre-
quency, has to be a function of location. For this reason,
a magnetic To excite a specific slice or volume at a specific location,
the MR resonance frequency, also called the Larmor fre-
quency, has to be a function of location. For this reason,
a magnetic field gradient is established in t

ing 23.2 MRI – System Components
Fig. 23.2a,b Commercially
available MR system using
a superconductive magnet ing 23.2 MRI – System Components

Fig. 23.2a, b Commercially

available MR system using

a superconductive magnet

with a magnetic field strength

of 3.0 T and patient hore ing 23.2 MRI – System Components 443

Fig. 23.2a, b Commercially

available MR system using

a superconductive magnet

with a magnetic field strength

of 3.0 T and patient bore

of 70cm (MAGNETOM ing 23.2 MRI – System Components

Fig. 23.2a, b Commercially

available MR system using

a superconductive magnet

with a magnetic field strength

of 3.0 T and patient bore

of 70 cm (MAGNETOM

SKYRA) The *T*_c weighted ing 23.2 MRI – System Components

Fig. 23.2a, b Commercially

available MR system using

a superconductive magnet

with a magnetic field strength

of 3.0 T and patient bore

of 70 cm (MAGNETOM

SKYRA). The T_1 -weighted
 ing 23.2 MRI – System Components

Fig. 23.2a, b Commercially

available MR system using

a superconductive magnet

with a magnetic field strength

of 3.0 T and patient bore

of 70 cm (MAGNETOM

SKYRA). The T_1 -weighted
 Fig. 23.2 MRI – System Components
 Fig. 23.2a, b Commercially

available MR system using

a superconductive magnet

with a magnetic field strength

of 3.0 T and patient bore

of 70 cm (MAGNETOM

SKYRA). The T_1 -weig **Fig. 23.2 MRI – System Components**
 Fig. 23.2a, b Commercially

available MR system using

a superconductive magnet

with a magnetic field strength

of 3.0 T and patient bore

of 70 cm (MAGNETOM

SKYRA). The T_1 -weig ing 23.2 MRI – System Components

Fig. 23.2a, b Commercially

available MR system using

a superconductive magnet

with a magnetic field strength

of 3.0 T and patient bore

of 70 cm (MAGNETOM

SKYRA). The T_1 -weighted
 Fig. 23.2a, b Commercially
available MR system using
a superconductive magnet
with a magnetic field strength
of 3.0 T and patient bore
of 70 cm (MAGNETOM
SKYRA). The T_1 -weighted
sagittal image of a human
brain demonstr Fig. 23.2a, b Commercially
available MR system using
a superconductive magnet
with a magnetic field strength
of 3.0 T and patient bore
of 70 cm (MAGNETOM
SKYRA). The T_1 -weighted
sagittal image of a human
brain demonstr

pradients that are linearly superimposed, providing the These forces cause vibration of the prossibility of producing any arbitrary excitation plane the primary source of noise during a without having to move any mechanica gradients that are linearly superimposed, providing the These forces cause vibration of possibility of producing any arbitrary excitation plane the primary source of noise due without having to move any mechanical parts. A **Example 19 The Supering Supering Supering Condition** and the proposibility of producing any arbitrary excitation plane the primary source of noise during without having to move any mechanical parts. A mag-
metic field gra gradients that are linearly superimposed, providing the These forces cause vibration possibility of producing any arbitrary excitation plane the primary source of noise dentited field gradient is also activated during the encoding. dients that are linearly superimposed, providing the These forces cause vibration of the sibility of producing any arbitrary excitation plane the primary source of noise during a coil gradient is also activated during the gradients that are linearly superimposed, providing the These forces cause vibration of the possibility of producing any arbitrary excitation plane the primary source of noise durine without having to move any mechanical possibility of producing any arbitrary excitation pl
without having to move any mechanical parts. A m
netic field gradient is also activated during the time
data acquisition. This encodes spatial information i
the signal

SKYRA). The T_1 -weighted
sagittal image of a human
brain demonstrates a slice
thickness of 1.2 mm, ac-
quired within 9 min
amplitude within 9 min
primary source of noise during an MR examination.
The amplitude of the ma sagittal image of a human
brain demonstrates a slice
thickness of 1.2 mm, ac-
quired within 9 min
according with respect to achieving an MR examination.
The amplitude of the magnetic field gradient is
important with respec brain demonstrates a slice
thickness of 1.2 mm, ac-
quired within 9 min
acquired within 9 min
These forces cause vibration of the gradient coil and are
the primary source of noise during an MR examination.
The amplitude of thickness of 1.2 mm, acquired within 9 min
quired within 9 min
These forces cause vibration of the gradient coil and are
the primary source of noise during an MR examination.
The amplitude of the magnetic field gradient is quired within 9 min

These forces cause vibration of the gradient coil and are

the primary source of noise during an MR examination.

The amplitude of the magnetic field gradient is

important with respect to achieving a These forces cause vibration of the gradient coil and are
the primary source of noise during an MR examination.
The amplitude of the magnetic field gradient is
important with respect to achieving a good spatial reso-
lutio These forces cause vibration of the gradient coil and are
the primary source of noise during an MR examination.
The amplitude of the magnetic field gradient is
important with respect to achieving a good spatial reso-
lutio These forces cause vibration of the gradient coil and are
the primary source of noise during an MR examination.
The amplitude of the magnetic field gradient is
important with respect to achieving a good spatial reso-
lutio These forces cause vibration of the gradient coil and are
the primary source of noise during an MR examination.
The amplitude of the magnetic field gradient is
important with respect to achieving a good spatial reso-
lutio These forces cause vibration of the gradient coil and are
the primary source of noise during an MR examination.
The amplitude of the magnetic field gradient is
important with respect to achieving a good spatial reso-
lutio the primary source of noise during an MR examination.
The amplitude of the magnetic field gradient is
important with respect to achieving a good spatial reso-
lution. A strong magnetic field gradient is also essential
for The amplitude of the magnetic field gradient is
important with respect to achieving a good spatial reso-
lution. A strong magnetic field gradient is also essential
for diffusion weighted imaging (DWI). Wherever the
amplit Magnetic Resonance Imaging [23,2 MRI – System Components
 Fig. 33.2a,b Components
 Fig. 33.2a,b Components
 Fig. 33.2a,b Components

as apperconductive magnet

with a magnetic field strength

of 3.0 T and patient ho

$$
F \approx I \times B_0.
$$

Fig. 23.3 Composition of ^a MR system

Part C | Medical Imaging

Air scaled and line to the mean and in 100 **Example 16.4**
Air-cooled gradient coils were used in 1986 with In case of knee examination, an explicit amplitudes of 3 mT/m and slew rate of ally used for excitation and signal
(ms). In 2003, generally available gra Medical Imaging
Air-cooled gradient coils were used in 1986 with In case of knee examination, an extraction amplitudes of 3 mT/m and slew rate of ally used for excitation and signal r
 3 T/(m s) . In 2003, generally a Medical Imaging

Air-cooled gradient coils were used in 1986 with In case of knee examination, an ex

gradient amplitudes of 3 mT/m and slew rate of ally used for excitation and signal
 3T/(m s) . In 2003, generally **Medical Imaging**

Air-cooled gradient coils were used in 1986 with In case of knee examination, an extr

gradient amplitudes of 3 mT/m and slew rate of ally used for excitation and signal $n \text{ sT/(m s)}$. In 2003, gener Medical Imaging

Air-cooled gradient coils were used in 1986 with In case of knee examination,

gradient amplitudes of 3 mT/m and slew rate of ally used for excitation and 3 T/(m s) . In 2003, generally available gra Medical Imaging
Air-cooled gradient coils were used in 1986 with In case of knee examination, an
gradient amplitudes of 3 mT/m and slew rate of ally used for excitation and sigt
 3 T/(m s) . In 2003, generally availab **Medical Imaging**
Air-cooled gradient coils were used in 1986 with In case of knee examination, an ext
gradient amplitudes of 3 mT/m and slew rate of ally used for excitation and signal 3 T/(m s) . In 2003, generally Medical Imaging

Air-cooled gradient coils were used in 1986 with In case of knee examination,

gradient amplitudes of 3 mT/m and slew rate of ally used for excitation and s

3 T/(m s). In 2003, generally available gradie **Medical Imaging**

Air-cooled gradient coils were used in 1986 with In case of knee examination, a

gradient amplitudes of 3 mT/m and slew rate of ally used for excitation and sig
 3 T/(m s) . In 2003, generally avai Air-cooled gradient coils were used in 1986 with In case of knee examination, gradient amplitudes of $3 \pi T/m$ and slew rate of ally used for excitation and $3 \text{ } T/(m \text{ s})$. In 2003, generally available gradient am-
plitude Air-cooled gradient coils were used in 1986 with In case of knee examination, gradient amplitudes of $3 \pi / (m s)$. In 2003, generally available gradient am-
only the slice within the kneed in $3 \text{ T} / (m \text{ m})$. In 2003, gene Air-cooled gradient coils were used in 1986 with In case of knee examination, an ex
gradient amplitudes of 3 mT/m and slew rate of ally used for excitation and signal
 3 T/(m s) . In 2003, generally available gradient gradient amplitudes of 3 mT/m and slew rate of ally used for excitation and signa 3T/(m s) . In 2003, generally available gradient am-
only the slice within the knee to plitudes were 45 mT/m with slew rate of 20 $3T/(m s)$. In 2003, generally available gradient am-
plitudes were 45 mT/m with slew rate of 200 T/(m s) vantage is that, in transverse ex
using water cooling. Ramp times to establish a magnetic knee is not excited, plitudes were 45 mT/m with slew rate of 200 T/(m s) vantage is that, in transverse e
using water cooling. Ramp times to establish a magnetic knee is not excited, which othe
field gradient are between 100 and 800 μ s, du using water cooling. Ramp times to establish a magnetic knee is not excited, which otherwis
field gradient are between 100 and 800 μ s, during which lead to imaging artifacts. In case
several hundred Amperes have to be field gradient are between 100 and 800 μ s, during which lead to imaging artifacts.
several hundred Amperes have to be maintained. To spine, large coil arrays, int
drive these currents, $\approx 2000 \text{ V}$ are needed. The de several hundred Amperes have to be maintained. To spine, large coil arrays, integrated
drive these currents, ≈ 2000 V are needed. The develop-
ment of faster and stronger gradient systems is slightly
limited by human p drive these currents, ≈ 2000 V are needed. The development of faster and stronger gradient systems is slightly limited by human physiology. Fast and rapid switching of magnetic field gradients induces electric potentia It of faster and stronger gradient systems is slightly
the d by human physiology. Fast and rapid switching
magnetic field gradients induces electric potentials
in the body that, with the current technology, can
the amplitu

it has to be avoided. All vendors providing strong
fast gradient systems have a *stimulation monitor*
changes imaging sequences prior to execution to a
peripheral nerve stimulation.
Beginning with the development of MR ima changes imaging sequences prior to execution to avoid
peripheral nerve stimulation.
Beginning with the development of MR imaging,
methods have been developed and established with the
aim of shortening measurement times by,

systems [23.19].
 23.2.3 The Radiofrequency System

The radiofrequency system consists of a transmit-

ter, transmit antenna, receive antenna, and receiver.

Depending on the magnetic field strength used, the

fundamenta **23.2.3 The Radiofrequency System**

The radiofrequency system consists of a transmit-

ter, transmit antenna, receive antenna, and receiver.

Depending on the magnetic field strength used, the

fundamental frequency will **23.2.3 The Radiofrequency System**

The radiofrequency system consists of a transmit-

ter, transmit antenna, receive antenna, and receiver.

Expeding on the magnetic field strength used, the

fundamental frequency will b The radiofrequency system consists of a transmit-
ter, transmit antenna, receive antenna, and receiver.
Depending on the magnetic field strength used, the
128 MHz (3.0T). The RF power amplifier has to have
a peak power of The radiofrequency system consists of a transmit-
ter, transmit antenna, receive antenna, and receiver.
Depending on the magnetic field strength used, the
fundamental frequency will be 8 MHz (0.2 T) up to
a peak power of s ter, transmit antenna, receive antenna, and receiver.

Depending on the magnetic field strength used, the

fundamental frequency will be 8 MHz (0.2 T) up to

128 MHz (3.0 T). The RF power amplifier has to have

of MR, a tr Depending on the magnetic field strength used, the
fundamental frequency will be 8 MHz (0.2 T) up to
128 MHz (3.0 T). The RF power amplifier has to have
of MR, a transmit antrenna, also called the body coil
is located imme fundamental frequency will be 8 MHz $(0.2$ T) up to 128 MHz $(3.0$ T). The RF power amplifier has to have a peak power of several kilowatts. Since the early days of MR, a transmit antenna, also called the body coil, t 128 MHz (3.0 T). The RF power amplifier has to have
a peak power of several kilowatts. Since the early days
of MR, a transmit antenna, also called the body coil,
is located immediately behind the cover within the pa-
the M a peak power of several kilowatts. Since the early days
of MR, a transmit antenna, also called the body coil,
is located immediately behind the cover within the pa-
tient bore. The body coil is usually also able to receive of MR, a transmit antenna, also called the body coil,
is located immediately behind the cover within the pa-
tient bore. The body coil is usually also able to receive
the MR signal. On the other hand, placement of a re-
ce is located immediately behind the cover within the pa-
tient bore. The body coil is usually also able to receive
the MR signal. On the other hand, placement of a re-
ceive antenna as close as possible to the patient's body tient bore. The body coil is usually also able to receive
the MR signal. On the other hand, placement of a re-
ceive antenna as close as possible to the patient's body
(surface coil) has some significant advantages. As the the MR signal. On the other hand, placement of a re-
ceive antenna as close as possible to the patient's body
(surface coil) has some significant advantages. As the
patient's body emits electromagnetic noise even in the
ab

In case of knee examination, an extremity coil is usu-
ally used for excitation and signal reception, exciting
only the slice within the knee to be studied. The ad-
vantage is that in transverse excitations the adjacent In case of knee examination, an extremity coil is usu-
ally used for excitation and signal reception, exciting
only the slice within the knee to be studied. The ad-
vantage is that, in transverse excitations, the adjacent
 In case of knee examination, an extremity coil is usu-
ally used for excitation and signal reception, exciting
only the slice within the knee to be studied. The ad-
vantage is that, in transverse excitations, the adjacent
 In case of knee examination, an extremity coil is usually used for excitation and signal reception, exciting
only the slice within the knee to be studied. The advantage is that, in transverse excitations, the adjacent
knee In case of knee examination, an extremity coil is usu-
ally used for excitation and signal reception, exciting
only the slice within the knee to be studied. The ad-
vantage is that, in transverse excitations, the adjacent
 In case of knee examination, an extremity coil is usually used for excitation and signal reception, exciting only the slice within the knee to be studied. The advantage is that, in transverse excitations, the adjacent knee In case of knee examination, an extremity coil is usually used for excitation and signal reception, exciting only the slice within the knee to be studied. The advantage is that, in transverse excitations, the adjacent knee

Magnetic Resonant

have been shown to be beneficial. Contrast-enhanced

MR angiography studies demand an even larger surface

coil coverage (Fig. 23.4) [23.20, 21]. Figure 23.5 shows

a coil arrangement suitable to cover t Magnetic Resonance Imaging

have been shown to be beneficial. Contrast-enhanced

MR angiography studies demand an even larger surface

coil coverage (Fig. 23.4) [23.20, 21]. Figure 23.5 shows

coil arrangement suitable to Magnetic Resonance

have been shown to be beneficial. Contrast-enhanced

MR angiography studies demand an even larger surface

coil coverage (Fig. 23.4) [23.20, 21]. Figure 23.5 shows

a coil arrangement suitable to cover Magnetic Resonance Imaging

Magnetic Resonance Imaging

MR angiography studies demand an even larger surface

coil coverage (Fig. 23.4) [23.20, 21]. Figure 23.5 shows

a coil arrangement suitable to cover to whole vascular M

Mave been shown to be beneficial. Contrast-enhanced

MR angiography studies demand an even larger surface

coil coverage (Fig. 23.4) [23.20, 21]. Figure 23.5 shows

a coil arrangement suitable to cover to whole vascular

MR angiography studies demand an even larger surface
coil coverage (Fig. 23.4) [23.20, 21]. Figure 23.5 shows
a coil arrangement suitable to cover to whole vascular
system from head to toe.
with the introduction of spatia coil coverage (Fig. 23.4) [23.20, 21]. Figure 23.5 shows
a coil arrangement suitable to cover to whole vascular
system from head to toe.
With the introduction of spatially distributed coil
arrays, some methods evolved to a coil arrangement suitable to cover to whole vascular
system from head to toe.
With the introduction of spatially distributed coil
arrays, some methods evolved to utilize the spatial in-
formation revolud by the coil dist system from head to toe.

With the introduction of spatially distributed coil

arrays, some methods evolved to utilize the spatial in-

formation provided by the coil distribution, reducing

receive the signal in parallel, With the introduction of spatially distributed coil
arrays, some methods evolved to utilize the spatial in-
formation provided by the coil distribution, reducing
the measurement time by undersampling. As all coils
sition t arrays, some methods evolved to utilize the spatial in-
formation provided by the coil distribution, reducing
the measurement time by undersampling. As all coils
receive the signal in parallel, the term parallel acqui-
 $\$ formation provided by the coil distribution, reducing
the measurement time by undersampling. As all coils
receive the signal in parallel, the term parallel acqui-
sition techniques (PAT) has been established, where the
de the measurement time by undersampling. As all coils
receive the signal in parallel, the term parallel acqui-
sition techniques (PAT) has been established, where the
expecting reduction in measurement time. The
undersampli receive the signal *in parallel*, the term *parallel acquisition techniques* (PAT) has been established, where the

PAT factor indicates the degree of undersampling and

the corresponding causes signal ambiguity leading t sition techniques (PAT) has been established, where the

PAT factor indicates the degree of undersampling and

the corresponding reduction in measurement time. The

undersampling causes signal ambiguity leading to so-

im PAT factor indicates the degree of undersampling and
the corresponding reduction in measurement time. The
undersampling causes signal ambiguity leading to so-
called *overfolding* artifacts. Prior to showing the final
the the corresponding reduction in measurement time. The

undersampling causes signal ambiguity leading to so-

called *overfolding* artifacts. Prior to showing the final

images to the user, a background task is to analyze
 undersampling causes signal ambiguity leading to so-
called *overfolding* artifacts. Prior to showing the final
images to the user, a background task is to analyze
the raw data or the final image of each coil to iden-
ang called *overfolding* artifacts. Prior to showing the final
images to the user, a background task is to analyze
the raw data or the final image of each coil to iden-
inty and remove such artifacts. The algorithm using
and images to the user, a background task is to analyze
the raw data or the final image of each coil to iden-
tify and remove such artifacts. The algorithm using
and peripheral angiography: head coil (1), necl
image informati

$$
SNR \sim \frac{1}{\sqrt{PAT}}.
$$

tems. Surement time with the spatial resolution being held

stant will result in a loss in SNR according to

This will lead to spatial inhomogen

SNR $\sim \frac{1}{\sqrt{PAT}}$.

I loss in SNR experienced when using parallel ac-

tives to constant will result in a loss in SNR according to
 $SNR \sim \frac{1}{\sqrt{PAT}}$.

SNR $\sim \frac{1}{\sqrt{PAT}}$.

The loss in SNR experienced when using parallel ac-

times to improve the underlying anatomy. The loss in SNR experienced when SNR $\sim \frac{1}{\sqrt{PAT}}$ homogeneous signal distributed to the underlying and
unwanted phenomenon. Besi
unwanted phenomenon. Besi
unwanted phenomenon. Besi
unwanted phenomenon. Besi
sixth achievable with higher magnetic field SNR $\sim \frac{1}{\sqrt{PAT}}$. The loss in SNR experienced when using parallel ac-

unwanted phenomenon. Besides

quisition techniques can be compensated with the better

still is distributed transmit coils can be under

For this r VPA1

unwanted phenomenon. Besides

unwanted phenomenon. Besides

quisition techniques can be compensated with the better

distributed transmit coils can be u

SNR achievable with higher magnetic field strength.

Besides The loss in SNR experienced when using parallel ac-

tives to compensate this artifici

quisition techniques can be compensated with the better

distributed transmit coils can be

SNR achievable with higher magnetic field

Following the term parallel acquisition techniques can be compensated with the better distributed transmit coils can
this reason, parallel imaging techniques have shown
full potential in conjunction with high-field sys-
 For this reason, parallel imaging techniques have shown
their full potential in conjunction with high-field sys-
tems.
Besides the utilization of spatially distributed sur-
face coils to improve the image quality or for t

their full potential in conjunction with high-field sys-
 and Image Reconstru

Erems.
 and Image Reconstru

frace coils to improve the image quality or for the At the beginning of the developp

purpose of reducing mea tems. **and Image Reconsti**

These coils to improve the image quality or for the At the beginning of the develop

purpose of reducing measurement time with undersam-

bulge purpose of reducing measurement time with undersa Besides the utilization of spatially distributed sur-
face coils to improve the image quality or for the At the beginning of the develop
purpose of reducing measurement time with undersam-
ing, documentation of patient da face coils to improve the image quality or for the At the beginning of the developp
purpose of reducing measurement time with undersam-
ing, documentation of patient data,
pling, it has been and still is discussed whether purpose of reducing measurement time with undersam-
pling, it has been and still is discussed whether it would
measurement protocols w
be of advantage to use spatially distributed coil arrays computer, and measurement
for pling, it has been and still is discussed whether it would
be of advantage to use spatially distributed coil arrays
computer, and measurement confor transmission, so-called transmit arrays (TX arrays). reconstruction were

sensitivity encoding (SENSE) [23.22]. Another algo-
rithm applied to the signal data with the same aim of B_0 is approaching the spatial dimens
removing the above-mentioned artifacts is called *gen*-body. This leads to rithm applied to the signal data with the same aim of B_0 is approaching the spatial dime

removing the above-mentioned artifacts is called *gen*-body. This leads to potential inter
 eralized autocalibrating partially removing the above-mentioned artifacts is called *gen* body. This leads to potential in
 eralized autocalibrating partially parallel acquisitions with and within the patient's b

(GRAPPA) [23.23]. All these PAT techniqu eralized autocalibrating partially parallel acquisitions with and within the patient's bod (GRAPPA) [23.23]. All these PAT techniques have one geneities. The latter results in disadvantage in common: Any attempt to reduce (GRAPPA) [23.23]. All these PAT techniques have one geneities. The latter results in diff
disadvantage in common: Any attempt to reduce the and/or refocusing amplitudes, leadin
measurement time with the spatial resolution **Fig. 23.5** Possible coil arrangement for a whole-body MR angiography: head coil (1), neck coil (2), body array (3), and peripheral angiography coil (4) (spine coil integrated within patient table)
 B_0 is approaching th **Fig. 23.5** Possible coil arrangement for a whole-body MR angiography: head coil (1), neck coil (2), body array (3), and peripheral angiography coil (4) (spine coil integrated within patient table)
 B_0 is approaching th **Fig. 23.5** Possible coil arrangement for a whole-body MR angiography: head coil (1), neck coil (2), body array (3), and peripheral angiography coil (4) (spine coil integrated within patient table)
 B_0 is approaching th **Fig. 23.5** Possible coil arrangement for a whole-body MR angiography: head coil (1), neck coil (2), body array (3), and peripheral angiography coil (4) (spine coil integrated within patient table)
 B_0 is approaching th **Fig. 23.5** Possible coil arrangement for a whole-body MR angiography: head coil (1), neck coil (2), body array (3), and peripheral angiography coil (4) (spine coil integrated within patient table)
 B_0 is approaching th angiography: head coil (1), neck coil (2), body array (3),
and peripheral angiography coil (4) (spine coil integrated
within patient table)
 B_0 is approaching the spatial dimensions of the patient's
body. This leads to and peripheral angiography coil (4) (spine coil integrated
within patient table)
 B_0 is approaching the spatial dimensions of the patient's
body. This leads to potential interaction of the EM field
with and within the p within patient table)
 B_0 is approaching the spatial dimensions of the patient's

body. This leads to potential interaction of the EM field

with and within the patient's body, causing B_1 inhomo-

geneities. The lat B_0 is approaching the spatial dimensions of the patient's
body. This leads to potential interaction of the EM field
with and within the patient's body, causing B_1 inhomo-
geneities. The latter results in different R B_0 is approaching the spatial dimensions of the patient's
body. This leads to potential interaction of the EM field
with and within the patient's body, causing B_1 inhomo-
geneities. The latter results in different R B_1 field. geneities. The latter results in different RF excitation
and/or refocusing amplitudes, leading to signal intensity
changes within the image that are unrelated to anatomy.
This will lead to spatial inhomogeneity of an othe focusing amplitudes, leading to signal intensity
within the image that are unrelated to anatomy.
lead to spatial inhomogeneity of an otherwise
eous signal distribution within the image, un-
the underlying anatomy. This is This will lead to spatial inhomogeneity of an otherwise
homogeneous signal distribution within the image, un-
related to the underlying anatomy. This is of course an
unwanted phenomenon. Besides other possible alterna-
ti homogeneous signal distribution within the image, un-
related to the underlying anatomy. This is of course an
unwanted phenomenon. Besides other possible alterna-
tives to compensate this artificial appearance, spatially

SNR achievable with higher magnetic field strength. B_1 field.
For this reason, parallel imaging techniques have shown
their full potential in conjunction with high-field sys-
the names and **Image Reconstruction**
tems.
 related to the underlying anatomy. This is of course an
unwanted phenomenon. Besides other possible alterna-
tives to compensate this artificial appearance, spatially
distributed transmit coils can be used to homogenize t momentum and phenomenon. Besides other possible alterna-
tives to compensate this artificial appearance, spatially
distributed transmit coils can be used to homogenize the
 B_1 field.
23.2.4 Measurement Control, Acquisi tives to compensate this artificial appearance, spatially
distributed transmit coils can be used to homogenize the
 B_1 field.
23.2.4 Measurement Control, Acquisition,
and **Image Reconstruction Systems**
At the beginnin distributed transmit coils can be used to homogenize the B_1 field.
 23.2.4 Measurement Control, Acquisition,
 and Image Reconstruction Systems

At the beginning of the development of MR imag-

img, documentation of **23.2.4 Measurement Control, Acquisition,**
 23.2.4 Measurement Control, Acquisition,
 23.2.4 Measurement Control, Acquisition
 24.1 The conclusion of patient data, image storage, and

measurement protocols were hand **23.2.4 Measurement Control, Acquisition,**
 and Image Reconstruction Systems

At the beginning of the development of MR imag-

ing, documentation of patient data, image storage, and

measurement protocols were handled b **23.2.4 Measurement Control, Acquisition,**

and **Image Reconstruction Systems**

At the beginning of the development of MR imag-

ing, documentation of patient data, image storage, and

measurement protocols were handled b **and Image Reconstruction Systems**
the beginning of the development of MR imag-
documentation of patient data, image storage, and
surement protocols were handled by a PDP-11
pputer, and measurement control as well as image ing, documentation of patient data, image storage, and
measurement protocols were handled by a PDP-11
computer, and measurement control as well as image
reconstruction were performed using vendor-specific
proprietary hardw

tems:

ages.

- ^A measurement control, providing the gradient am-**Example 18 Exercise 19 Exerci Example 18 I maging**
 Example 18 I maging the senerated im-
 Example 18 I maginal information is contain

ages.

A measurement control, providing the gradient am-

plifier with data necessary to establish the magnetic **Solution**
 Solution
 Solution
 Solution
 Example 1998
 Example 1999
 Example 1999 Example 15 The synchronization the measurements, and displaying the generated im-

A measurement control, providing the gradient am-

MR signal. That in

plifier with data necessary to establish the magnetic

gradient fi system. **Example 18 An image and the phase information as compared and the phase information of An image reconstruction plifier with data necessary to establish the magnetic a Fourie** the measurements, and displaying the generated im-
ages. we separated measurement control, providing the gradient am-
A *measurement control*, providing the gradient am-
plifier with data necessary to establish the magneti • A *measurement control*, providing the gradient am-

plifier with data necessary to establish the magnetic a Fourier transformation. See

gradient fields on time, supplying the RF power am-

plifier with the necessary i gradient fields on time, supplying the RF power am-

image reconstruction system

plifier with the necessary information, and handling

ing algorithms that run pri

the synchronization with the image reconstruction, or im
-

Example 16.4
 the measurements, and displaying the generated im-

ages.
 A measurement control, providing the gradient am-
 A measurement control, providing the gradient am-
 A Fourier transformation
 A Fourier spatial information is contained within the frequency and the phase information of the received
MR signal. That information is retriveed using
a Fourier transformation. Secondary tasks for the spatial information is contained within the frequency and the phase information of the received
MR signal. That information is retrieved using
a Fourier transformation. Secondary tasks for the
image reconstruction system a spatial information is contained within the frequency and the phase information of the received MR signal. That information is retrieved using a Fourier transformation. Secondary tasks for the propreses-
image reconstructi spatial information is contained within the frequency and the phase information of the received MR signal. That information is retrieved using a Fourier transformation. Secondary tasks for the image reconstruction system a spatial information is contained within the frequency and the phase information of the received MR signal. That information is retrieved using a Fourier transformation. Secondary tasks for the image reconstruction system a spatial information is contained within the frequency and the phase information of the received MR signal. That information is retrieved using a Fourier transformation. Secondary tasks for the image reconstruction system a spatial information is contained within the frequency and the phase information of the received MR signal. That information is retrieved using a Fourier transformation. Secondary tasks for the image reconstruction system a spatial information is contained within the frequency and the phase information of the received MR signal. That information is retrieved using a Fourier transformation. Secondary tasks for the image reconstruction system a spatial information is contained within the frequency and the phase information of the received MR signal. That information is retrieved using a Fourier transformation. Secondary tasks for the image reconstruction system a

ifter with the necessary information, and handling

ing algorithms that run prior

the synchronization with the image reconstruction or immediately

system.
 • An *image reconstruction system* with the primary In theory the synchronization with the image reconstruction reconstruction, or immediately
system.

• An *image reconstruction system* with the primary In theory it is also conceivable to

task of performing fast Fourier transforma system.

• An *image reconstruction system* with the primary In theory it is also conceivable

task of performing fast Fourier transformation. The tasks within one computer [23.2
 23.3 MRI – Basic Principles and Applicat • An *image reconstruction system* with the primary In theory it is also conceivat task of performing fast Fourier transformation. The tasks within one computer [2
 23.3 MRI – **Basic Principles and Applications**

MRI is task of performing fast Fourier transformation. The tasks within one computer [2
 23.3 MRI – **Basic Principles and Applications**

MRI is a well-established modality for routine clinical ical imaging. The basis for

imag **23.3 MRI – Basic Principles and Applications**
MRI is a well-established modality for routine clinical ical imaging. The basis for a
imaging. In conjunction with healthcare reform, there is the essential series of action
 23.3 MRI – Basic Principles and Applications

MRI is a well-established modality for routine clinical ical imaging. The basis fo

imaging. In conjunction with healthcare reform, there is the essential series of a

are s **23.3 MRI – Basic Principles and Applications**

MRI is a well-established modality for routine clinical ical imaging. The basis for imaging. In conjunction with healthcare reform, there is the essential series of action a **23.3 MRI – Basic Principles and Applications**

MRI is a well-established modality for routine clinical ical imaging. The basis for

imaging. In conjunction with healthcare reform, there is the essential series of ac

are MRI is a well-established modality for routine clinical
inaging. The basis for a me
imaging. In conjunction with healthcare reform, there is the essential series of actions f
are standards and guidelines describing as and MRI is a well-established modality for routine clinical
inaging. The basis for
imaging. In conjunction with healthcare reform, there
is the essential series of act
are standards and guidelines describing as and when cusin imaging. In conjunction with healthcare reform, there is the essential series of actions for are standards and guidelines describing as and when cusing, spatial encoding, and data a MRI is appropriate [23.25, 26]. To ensu are standards and guidelines describing as and when

MRI is appropriate [23.25, 26]. To ensure the quality

sequence.

of diagnostic studies, these standards and guidelines

are supported by matching billing codes for rei MRI is appropriate [23.25, 26]. To ensure the quality sequence.

of diagnostic studies, these standards and guidelines

are supported by matching billing codes for reimburse-
 23.3.1 Slice Selection and Sp

ment [23.27] of diagnostic studies, these standards and guidelines
are supported by matching billing codes for reimburse-
meansure constantly implementing cost-savings approaches
The sequence starts with ramp
in healthcare to ensure a are supported by matching billing codes for reimburse-
ment [23.27]. Medicare and Medicaid service centers
are constantly implementing cost-savings approaches
The sequence starts with ramping in
healthcare tystem [23.28]. ment [23.27]. Medicare and Medicaid service centers
are constantly implementing cost-savings approaches
in healthcare to ensure an affordable and adequate field gradient in the direction of s
healthcare system [23.28]. St are constantly implementing cost-savings approaches The sequence starts with rar
in healthcare to ensure an affordable and adequate field gradient in the directi
healthcare system [23.28]. Standards and guidelines shown i in healthcare to ensure an affordable and adequate field gradient in the direction
healthcare system [23.28]. Standards and guidelines shown in Fig. 23.6. As soon as t
include recommendations with respect to spatial res-
 healthcare system [23.28]. Standards and guidelines shown in Fig. 23.6. As soon as the include recommendations with respect to spatial res-
include, a RF pulse is applie olution, weightings, slice orientations, and covera include recommendations with respect to spatial res-
olution, weightings, slice orientations, and coverage. Transport and the Larmot
The nomenclature *weighting* is used to indicate the gion to be excited. A use
tissue-sp olution, weightings, slice orientations, and coverage. Tange matching the Larmor fre
The nomenclature *weighting* is used to indicate the gion to be excited. A user-det
tissue-specific MR-related parameter that dominates The nomenclature weighting is used to indicate the gion to be excited. A user-defined
tissue-specific MR-related parameter that dominates the matrix define the size of a single spa
image contrast. The sources for MR signa tissue-specific MR-related parameter that dominates the matrix define the size of a
image contrast. The sources for MR signals are pro-
a voxel. The sum of all s
tons (primarily the nuclei of hydrogen atoms located gle vo image contrast. The sources for MR signals are pro-
the sum of all signos (primarily the nuclei of hydrogen atoms located gle voxel defines the bright
within relatively freely moving water molecules). If the pixel on the tons (primarily the nuclei of hydrogen atoms located gle voxel defines the brightness c
within relatively freely moving water molecules). If the pixel on the screen. The only pher
imaging protocol is selected such that th within relatively freely moving water molecules). If the pixel on the screen. The only phe
imaging protocol is selected such that the number of spatial encoding is the fact that
protons within a voxel dominates the signal imaging protocol is selected such that the number of spatial encoding is the fact
protons within a voxel dominates the signal amplitude quency of the rotating transv
and thereby the image contrast, the weighting is called protons within a voxel dominates the signal amplitude
and thereby the image contrast, the weighting is called is a function of the magneti
proton density (PDw). If the factor dominating the im-
ion. If a brief period of d and thereby the image contrast, the weighting is called is a function of the magnetic field s
proton density (PDw). If the factor dominating the im-
ion. If a brief period of different ro
age contrast is the speed of reco proton density (PDw). If the factor dominating the im-

ion. If a brief period of different ro

age contrast is the speed of recovery of the longitudinal

is utilized, the phase position of the

nuclear magnetization foll age contrast is the speed of recovery of the longitudinal is utilized, the phase positionuclear magnetization following excitation, the weight-
inclear magnetizations wi
ing is called T_1w . If the image contrast is domi nuclear magnetization following excitation, the weight-
nuclear magnetizations will be
ing is called T_1w . If the image contrast is dominated by
phase encoding. Such phase ence
the effect of the tissue-specific fading o ing is called T_1w . If the image contrast is dominated by
the effect of the tissue-specific fading of the MR sig-
all, the weighting is called T_2w . Based on experience
the direction of slice selection. *A*
acquired o the effect of the tissue-specific fading of the MR sig-

inctuded mal, the weighting is called T_2w . Based on experience

the direction of slice selection. A

acquired over decades, vendors provide programs con-

time t and is stored in a raw data matrix. As each data point Part C 23.3

image reconstruction system are inline postprocess-
ing algorithms that run prior to or during image
reconstruction, or immediately afterwards.
In theory it is also conceivable to combine all of these
tasks within one comp ing algorithms that run prior to or during image
reconstruction, or immediately afterwards.
In theory it is also conceivable to combine all of these
tasks within one computer [23.24].
Cations
Cations
ical imaging. The reconstruction, or immediately afterwards.

In theory it is also conceivable to combine all of these

tasks within one computer [23.24].
 Cations

ical imaging. The basis for a measurement protocol

is the essential seri **Cations**
 C Combinding Example 15

ical imaging. The basis for a measurement protocol

is the essential series of actions for excitation, refo-
 23.3.1 Slice Selection and Spatial Encoding
 The sequence starts with ramping up of

Cations

ical imaging. The basis for a measurement protocol

is the essential series of actions for excitation, refo-

cusing, spatial encoding, and data acquisition, called a
 23.3.1 Slice Selection and Spatial Encodin Cations

ical imaging. The basis for a measurement protocol

is the essential series of actions for excitation, refo-

cusing, spatial encoding, and data acquisition, called a
 23.3.1 Slice Selection and Spatial Encodi ical imaging. The basis for a measurement protocol
is the essential series of actions for excitation, refo-
cusing, spatial encoding, and data acquisition, called a
23.3.1 Slice Selection and Spatial Encoding
The sequenc ical imaging. The basis for a measurement protocol
is the essential series of actions for excitation, refo-
cusing, spatial encoding, and data acquisition, called a
sequence.
23.3.1 Slice Selection and Spatial Encoding is the essential series of actions for excitation, refocusing, spatial encoding, and data acquisition, called a
sequence.
23.3.1 Slice Selection and Spatial Encoding
The sequence starts with ramping up of the magnetic
f cusing, spatial encoding, and data acquisition, called a

sequence.
 23.3.1 Slice Selection and Spatial Encoding

The sequence starts with ramping up of the magnetic

field gradient in the direction of slice selection, a **23.3.1 Slice Selection and Spatial Encoding**
 23.3.1 Slice Selection and Spatial Encoding

The sequence starts with ramping up of the magnetic

field gradient in the direction of slice selection, as

shown in Fig. 23.6 **23.3.1 Slice Selection and Spatial Encoding**
The sequence starts with ramping up of the magnetic
field gradient in the direction of slice selection, as
shown in Fig. 23.6. As soon as the magnetic field gra-
dient is stab **23.3.1 Slice Selection and Spatial Encoding**
The sequence starts with ramping up of the magnetic
field gradient in the direction of slice selection, as
shown in Fig. 23.6. As soon as the magnetic field gra-
dient is stab The sequence starts with ramping up of the magnetic
field gradient in the direction of slice selection, as
shown in Fig. 23.6. As soon as the magnetic field gra-
dient is stable, a RF pulse is applied with a frequency
rang The sequence starts with ramping up of the magnetic
field gradient in the direction of slice selection, as
shown in Fig. 23.6. As soon as the magnetic field gra-
dient is stable, a RF pulse is applied with a frequency
rang field gradient in the direction of slice selection, as
shown in Fig. 23.6. As soon as the magnetic field gra-
dient is stable, a RF pulse is applied with a frequency
range matching the Larmor frequencies within the re-
gio shown in Fig. 23.6. As soon as the magnetic field gradient is stable, a RF pulse is applied with a frequency
range matching the Larmor frequencies within the re-
gion to be excited. A user-defined field of view and
matrix dient is stable, a RF pulse is applied with a frequency
range matching the Larmor frequencies within the re-
gion to be excited. A user-defined field of view and
matrix define the size of a single spatial volume, called
a range matching the Larmor frequencies within the region to be excited. A user-defined field of view and matrix define the size of a single spatial volume, called a *voxel*. The sum of all signals coming out of a single vox gion to be excited. A user-defined field of view and
matrix define the size of a single spatial volume, called
a *voxel*. The sum of all signals coming out of a sin-
gle voxel defines the brightness of the corresponding
pi matrix define the size of a single spatial volume, called
a *voxel*. The sum of all signals coming out of a sin-
gle voxel defines the brightness of the corresponding
pixel on the screen. The only phenomenon utilized for
s a *voxel*. The sum of all signals coming out of a sin-
gle voxel defines the brightness of the corresponding
pixel on the screen. The only phenomenon utilized for
spatial encoding is the fact that the precessional fre-
que gle voxel defines the brightness of the corresponding
pixel on the screen. The only phenomenon utilized for
spatial encoding is the fact that the precessional fre-
quency of the rotating transverse nuclear magnetization
is pixel on the screen. The only phenomenon utilized for spatial encoding is the fact that the precessional frequency of the rotating transverse nuclear magnetization is a function of the magnetic field strength at that loca spatial encoding is the fact that the precessional frequency of the rotating transverse nuclear magnetization
is a function of the magnetic field strength at that loca-
tion. If a brief period of different rotational frequ quency of the rotating transverse nuclear magnetization
is a function of the magnetic field strength at that loca-
tion. If a brief period of different rotational frequencies
is utilized, the phase position of the adjacent is a function of the magnetic field strength at that location. If a brief period of different rotational frequencies
is utilized, the phase position of the adjacent transverse
nuclear magnetizations will be altered. This tion. If a brief period of different rotational frequencies
is utilized, the phase position of the adjacent transverse
nuclear magnetizations will be altered. This is called
phase encoding. Such phase encoding is usually is utilized, the phase position of the adjacent transverse
nuclear magnetizations will be altered. This is called
phase encoding. Such phase encoding is usually applied
directly after excitation in a direction perpendicula nuclear magnetizations will be altered. This is called
phase encoding. Such phase encoding is usually applied
directly after excitation in a direction perpendicular to
the direction of slice selection. At the same point in phase encoding. Such phase encoding is usually applied
directly after excitation in a direction perpendicular to
the direction of slice selection. At the same point in
time the expected dephasing during frequency encod-
in

THE MR signal

THE 23.6 Sequence diagram. A RF excitation pulse is applied as soon as the slice selection gradient (GS) reaches the

interior of slice selection dramatic material but (due to the differences in resonance f **EFT**
 ERECT:
 ERE GA
 a consequence of the frequency-encoding magnetic field gradient (GA) in the first consequence of the frequencies in the direction of slice selection during the excitation pulse, the generated but (due to the differe **in the presence of a** frequency-encoding magnetic field gradient (GA). The data are saved into a raw data matrix, also diel are presence of slice selection provides the mominal value. Following the excitation pulse, the GA
 Fig. 23.6 Sequence diagram. A RF excitation pulse is applied as soon as the slice selection gradient (GS) reaches the

nominal value. Following the excitation pulse, the generated but (due to the differences in reso **Fig. 23.6** Sequence diagram. A RF excitation pulse is applied as soon as the slice selection gradition
nominal value. Following the excitation pulse, the generated but (due to the differences in resonance
direction of sl **Fig. 23.6** Sequence diagram. A RF excitation pulse is applied as soon as the slice selection gradie
nominal value. Following the excitation pulse, the generated but (due to the differences in resonance
direction of slice nominal value. Following the excitation pulse, the generated but (due to the differences in resonardirection of slice selection during excitation) dephased transverse nuclear magnetization is rephatime period the phase-en direction of slice selection during excitation) dephased transverse nuclear magnetization is it
ime period the phase-encoding gradient can be applied (GP) and the rephasing in the direction
be prepared (GA) – as the trans time period the phase-encoding gradient can be applied (GP) and the rephasing in the direction
be prepared (GA) – as the transverse nuclear magnetization will dephase due to the difference
a consequence of the frequency-e d as soon as the slice selection gradient (GS) reaches the
but (due to the differences in resonance frequencies in the
verse nuclear magnetization is rephased. During the same
and the rephasing in the direction of frequenc d as soon as the slice selection gradient (GS) reaches the
but (due to the differences in resonance frequencies in the
verse nuclear magnetization is rephased. During the same
and the rephasing in the direction of frequenc but (due to the differences in resonance frequencies in the verse nuclear magnetization is rephased. During the same and the rephasing in the direction of frequency encoding can II dephase due to the differences in resona werse nuclear magnetization is rephased. During the same
nd the rephasing in the direction of frequency encoding can
ll dephase due to the differences in resonance frequencies as
ent during the readout period. Finally, th the rephasing in the direction of frequency encoding
ephase due to the differences in resonance frequencies
during the readout period. Finally, the data are acqui
(GA). The data are saved into a raw data matrix, a
FT) app

be prepared (GA) – as the transverse nuclear magnum a consequence of the frequency-encoding magne
in the presence of a frequency-encoding magne
called *k*-space. A two-dimensional fast Fourier t
within the raw data matri in the presence of a frequency-encoding magnetic field gradient (GA). T
called *k*-space. A two-dimensional fast Fourier transformation (FFT) appl
within the raw data matrix has an index called *k*, the variations or
aw d within the raw data matrix has an index called k, the

raw data matrix is also called k-*space*. Depending on

tial dimensions of a voxel will cause

the matrix size in the direction of phase encoding, mul-

frequencies, within the raw data matrix has an index called k, the variations of the
raw data matrix is also called k-space. Depending on tial dimensions of
the matrix size in the direction of phase encoding, mul-
frequencies, resurvi within the raw data matrix has an index called k, the variations of the magnetic sus
raw data matrix is also called k-space. Depending on tial dimensions of a voxel will
the matrix size in the direction of phase encoding,

Fracta matrix is also called *k-space*. Depending on tial dimensions of a voxel will cause
the matrix size in the direction of phase encoding, mul-
frequencies, resulting in faster fadin,
tiple repetitions with different the matrix size in the direction of phase encoding, mul-

frequencies, resulting in faster fading

tiple repetitions with different phase-encoding gradients

This faster fading is characterized by

are required to obtain tiple repetitions with different phase-encoding gradients This faster fading is characteriz
are required to obtain enough information to reconstruct T_2^* , given by
an image.
23.3.2 The Spin-Echo Sequence with γ be are required to obtain enough information to reconstruct T_2^* , given by

an image.
 $\frac{1}{T_2^*} = \frac{1}{T_2} + \gamma \Delta B$,

23.3.2 The Spin-Echo Sequence

The spin echo was found accidentally during an ex-

with γ being the **23.3.2 The Spin-Echo Sequence**

T^{*}/₂ = $\frac{1}{T_2}$ + $\gamma \Delta B$,

The spin echo was found accidentally during an ex-

periment to measure a tissue-specific T_1 relaxation magnetic field gradient cause

time [23.29]. C **23.3.2 The Spin-Echo Sequence**

The spin echo was found accidentally during an ex-

T⁻¹ s⁻¹ for the proto

periment to measure a tissue-specific T_1 relaxation magnetic field gradient

time [23.29]. Combined with t The spin echo was found accidentally during an ex-

with γ being the gyrom

periment to measure a tissue-specific T_1 relaxation magnetic field gradient (time [23.29]. Combined with the spatial encoding magnetic susc behavior). ment to measure a tissue-specific T_1 relaxation magnetic field gradient caused
 \geq [23.29]. Combined with the spatial encoding magnetic susceptibility. The res

eme of an imaging sequence it is also termed the tion time [23.29]. Combined with the spatial encoding magnetic susceptibility. T
scheme of an imaging sequence it is also termed the tion of different resonance
spin-echo sequence. Initially introduced for imaging in and const 1983, it is still used today for T_1w imaging. Besides

the already discussed tissue-specific parameters PD, T_1 , echo sequence has a 90° RF exci

and T_2 , there is another imaging-relevant parameter, a 180° RF refo the already discussed tissue-specific parameters PD, T_1 , echo sequence has a 90° RF e
and T_2 , there is another imaging-relevant parameter, a 180° RF refocusing pulse. We
the magnetic susceptibility χ , which indic and T_2 , there is another imaging-relevant parameter, a 180° RF refocusing
the magnetic susceptibility χ , which indicates whether cusing pulse, the fast
the external magnetic field is increased (paramagnetic nuclear

$$
B_{0(\text{eff})} = (1 + \chi)B_0.
$$

the magnetic susceptibility χ , which indicates whether

the external magnetic field is increased (paramagnetic

or ferromagnetic behavior) or decreased (diamagnetic

component, and in the process obtavior).

With this,

 T_2^* , given by 1 (FFT) applied to those data will lead to the final image
variations of the magnetic susceptibility within the spa-
tial dimensions of a voxel will cause different resonance
frequencies, resulting in faster fading of th for the magnetic susceptibility within the spa-
ions of a voxel will cause different resonance
s, resulting in faster fading of the MR signal.
fading is characterized by the time constant
by
 $\frac{1}{T_2} + \gamma \Delta B$,
ing the gyr

$$
\frac{1}{T_2^*} = \frac{1}{T_2} + \gamma \Delta B ,
$$

 T^{-1} s⁻¹ for the proton, a variations of the magnetic susceptibility within the spa-
tial dimensions of a voxel will cause different resonance
frequencies, resulting in faster fading of the MR signal.
This faster fading is characterized by the time variations of the magnetic susceptibility within the spa-
tial dimensions of a voxel will cause different resonance
frequencies, resulting in faster fading of the MR signal.
This faster fading is characterized by the time tial dimensions of a voxel will cause different resonance
frequencies, resulting in faster fading of the MR signal.
This faster fading is characterized by the time constant
 T_2^* , given by
 $\frac{1}{T_2^*} = \frac{1}{T_2} + \gamma \Delta B$, frequencies, resulting in faster fading of the MR signal.
This faster fading is characterized by the time constant
 T_2^* , given by
 $\frac{1}{T_2^*} = \frac{1}{T_2} + \gamma \Delta B$,
with γ being the gyromagnetic ratio of 2.675 × 10⁸
This faster fading is characterized by the time constant T_2^* , given by
 $\frac{1}{T_2^*} = \frac{1}{T_2} + \gamma \Delta B$,

with γ being the gyromagnetic ratio of 2.675×10^8
 T^{-1} s⁻¹ for the proton, and ΔB representing the T_2^* , given by
 $\frac{1}{T_2^*} = \frac{1}{T_2} + \gamma \Delta B$,

with γ being the gyromagnetic ratio of 2.675×10⁸
 T^{-1} s⁻¹ for the proton, and ΔB representing the

magnetic field gradient caused by the difference in

magne $\frac{1}{T_2^*} = \frac{1}{T_2} + \gamma \Delta B$,

with γ being the gyromagnetic ratio of 2.675×10⁸

T⁻¹ s⁻¹ for the proton, and ΔB representing the

magnetic field gradient caused by the difference in

magnetic susceptibility. $\frac{1}{T_2^*} = \frac{1}{T_2} + \gamma \Delta B$,
with γ being the gyromagnetic ratio of 2.675 × 10⁸
T⁻¹ s⁻¹ for the proton, and ΔB representing the
magnetic field gradient caused by the difference in
magnetic susceptibility. Th ²

²

with γ being the gyromagnetic ratio of 2.675×10⁸

T⁻¹ s⁻¹ for the proton, and ΔB representing the

magnetic field gradient caused by the difference in

magnetic susceptibility. The resulting spatial with γ being the gyromagnetic ratio of 2.675×10^8
 T^{-1} s⁻¹ for the proton, and ΔB representing the

magnetic field gradient caused by the difference in

magnetic susceptibility. The resulting spatial distri T^{-1} s⁻¹ for the proton, and ΔB representing the
magnetic field gradient caused by the difference in
magnetic susceptibility. The resulting spatial distribu-
tion of different resonance frequencies is spatially fix magnetic field gradient caused by the difference in
magnetic susceptibility. The resulting spatial distribu-
tion of different resonance frequencies is spatially fixed
and constant in time. This allows the dephasing to be
 gnetic susceptibility. The resulting spatial distribu-
of different resonance frequencies is spatially fixed
constant in time. This allows the dephasing to be
resed using a so-called RF refocusing pulse. A spin-
o sequenc tion of different resonance frequencies is spatially fixed
and constant in time. This allows the dephasing to be
reversed using a so-called RF refocusing pulse. A spin-
echo sequence has a 90° RF excitation pulse and also and constant in time. This allows the dephasing to be
reversed using a so-called RF refocusing pulse. A spin-
echo sequence has a 90° RF excitation pulse and also
a 180° RF refocusing pulse. With this 180° RF refo-
cusing reversed using a so-called RF refocusing pulse. A spin-
echo sequence has a 90° RF excitation pulse and also
a 180° RF refocusing pulse. With this 180° RF refo-
cusing pulse, the faster component of the transverse
nuclear Resonance Imaging [23.3 MRI – Basic Principles and Applications 447

and as soon as the slice sidection gradient (GS) readeds the

between molecular time is preased theorem in the properties in the

and the replacing in t

parameter, a 180° RF refocusing pulse. With
tes whether cusing pulse, the faster component
iramagnetic nuclear magnetization will be placee
iliamagnetic component, and in the process of cecho is formed, which is acqu the external magnetic field is increased (paramagnetic nuclear magnetization will be place
or ferromagnetic behavior) or decreased (diamagnetic component, and in the process of ceho
is formed, which is acquired
with this, echo sequence has a 90° RF excitation pulse and also
a 180° RF refocusing pulse. With this 180° RF refo-
cusing pulse, the faster component of the transverse
nuclear magnetization will be placed behind the slower
componen a 180° RF refocusing pulse. With this 180° RF refocusing pulse, the faster component of the transverse nuclear magnetization will be placed behind the slower component, and in the process of catching up, a spin echo is fo cusing pulse, the faster component of the transverse
nuclear magnetization will be placed behind the slower
component, and in the process of catching up, a spin
echo is formed, which is acquired in the presence of
a frequ

Fig. 23.7 Sequence diagram for
a spin-echo sequence, illustrating the
repetition time (TR) and echo time Fig. 23.7 Sequence diagram for
a spin-echo sequence, illustrating the
repetition time (TR) and echo time
(TE) Fig. 23.7 Sequence diagram for
a spin-echo sequence, illustrating the
repetition time (TR) and echo time
(TE) (TE)

echo time will lead to a ^T2-weighted (^T2^w) protocol. Repetition time TR

Repetition times will be suppressed as of the relaxation times with

well. Such a protocol is called proton density weighting ble 23.1). The matrix size i

(PDw), as primarily the number of protons per **Example 18**
 Example 18 Transfer Constant Constant Constant Constant Constant Constant Constant Constant Constant Cell. Such a protocol is called proton density weighting be 23.1). The matrix size in the (PDw), as prima Example 18 Example 18 Example 18
ferences in T_2 relaxation times will be suppressed as of the relaxation times within bi
well. Such a protocol is called proton density weighting ble 23.1). The matrix size in the
(PDw ferences in T_2 relaxation times will be suppressed as of the relaxation times w
well. Such a protocol is called proton density weighting ble 23.1). The matrix siz
(PDw), as primarily the number of protons per voxel enc

ity weighting ble 23.1). The matrix size in the direction of phase
ms per voxel encoding usually dictates the number of necessary rep-
eMR signal. etitions and with this the overall measurement time.
ing a longer For a T of the relaxation times within biological tissues (Ta-
ble 23.1). The matrix size in the direction of phase
ercoding usually dictates the number of necessary rep-
ertitions and with this the overall measurement time of the relaxation times within biological tissues (Ta-
ble 23.1). The matrix size in the direction of phase
encoding usually dictates the number of necessary rep-
etitions and with this the overall measurement time.
For a encoding usually dictates the number of necessary rep-
the 23.1). The matrix size in the direction of phase
encoding usually dictates the number of necessary rep-
etitions and with this the overall measurement time.
For a etime the relaxation times within biological tissues (Ta-
ble 23.1). The matrix size in the direction of phase
encoding usually dictates the number of necessary rep-
etitions and with this the overall measurement time.
For For the relaxation times within biological tissues (Ta-
ble 23.1). The matrix size in the direction of phase
encoding usually dictates the number of necessary rep-
etitions and with this the overall measurement time.
For a the relaxation times within biological tissues (Ta-
ble 23.1). The matrix size in the direction of phase
encoding usually dictates the number of necessary rep-
etitions and with this the overall measurement time.
For a T excitation immediates within biological tissues (Table 23.1). The matrix size in the direction of phase encoding usually dictates the number of necessary repetitions and with this the overall measurement time.
For a T_1 shifted and times within biological tissues (Ta-
ble 23.1). The matrix size in the direction of phase
encoding usually dictates the number of necessary rep-
etitions and with this the overall measurement time.
For a T_1

Magnetic Resonance Imaging 23.3 MRI – Basic
ment, the surrounding tissue. Aqueous solutions such as multiple RF refocusing pulses. On
ccrebrospinal fluid show long T_1 relaxation time and ap-
pear hypointens (dark) on Magnetic Resonance Imaging 23.3 MRI – Basi
ment, the surrounding tissue. Aqueous solutions such as multiple RF refocusing pulses. On
cerebrospinal fluid show long T₁ relaxation time and ap-
heasions such as the surrors o Magnetic Resonance Imaging 23.3 MRI – Bannon
ment, the surrounding tissue. Aqueous solutions such as multiple RF refocusing pulses. C
cerebrospinal fluid show long T_1 relaxation time and ap-
pear hypointense (dark) on Magnetic Resonance Imaging 23.3 MRI

ment, the surrounding tissue. Aqueous solutions such as multiple RF refocusing pulse

cerebrospinal fluid show long T_1 relaxation time and ap-

pear hypointense (dark) on T_1 -weig Magnetic Resonance Imaging 23.3 MRI –
ment, the surrounding tissue. Aqueous solutions such as multiple RF refocusing pulses.
cerebrospinal fluid show long T_1 relaxation time and apages, it becomes obvious that
pear hyp Magnetic Resonance Imaging 23.3 MRI – Ba

ment, the surrounding tissue. Aqueous solutions such as multiple RF refocusing pulses. O

ccrebrospinal fluid show long T_1 relaxation time and appearance (dark) on T_1 - weigh Magnetic Resonance Imaging 23.3 MRI – Batter Resonance Imaging 23.3 MRI – Batter

ment, the surrounding tissue. Aqueous solutions such as multiple RF refocusing pulses. O

cerebrospinal fluid show long T_1 relaxation ti Magnetic Resonance Imaging 23.3

ment, the surrounding tissue. Aqueous solutions such as

multiple RF refocusing

cerebrospinal fluid show long T_1 relaxation time and ap-

opear hypointense (dark) on T_1 -weighted ima ment, the surrounding tissue. Aqueous solutions such as

ecerebrospinal fluid show long T_1 relaxation time and ap-

pear hypointense (dark) on T_1 -weighted images. Mass

lightly changing, especially the

lesions such ment, the surrounding tissue. Aqueous solutions such as

cerebrospinal fluid show long T_1 relaxation time and ap-

ages, it becomes obvious that im

pear hypointense (dark) on T_1 -weighted images. Mass slightly chang ment, the surrounding tissue. Aqueous solutions such as

multiple RF refocusing

cerebrospinal fluid show long T_1 relaxation time and ap-

ages, it becomes obvide

pear hypointense (dark) on T_1 -weighted images. Mass cerebrospinal fluid show long T_1 relaxation time and ap-
pear hypointense (dark) on T_1 -weighted images. Mass slightly changing, especially for im
lesions such as tumors often not only show displace-
late echoes. Thi pear hypointense (dark) on T_1 -weighted images. Mass slightly changing, especial
esions such as tumors often not only show displace-
ment of normal anatomy but usually involve edema, another k-space line rathed
ocumente weighted images. With a few exceptions, tissues with the measurement time (Fig. 23.1

long T_1 relaxation times usually also demonstrate long
 T_2 relaxation times, as a consequence of intramolecular

where molecules. long T_1 relaxation times usually also demonstrate long
 T_2 relaxation times, as a consequence of intramolecular

dipole–dipole interactions within the rapidly tumbling

water molecules. For this reason, pathologic t T₂ relaxation times, as a consequence of intramolecular acronym *rapid acquisition*
dipole–dipole interactions within the rapidly tumbling *ment* (RARE) [23.34]. The in
water molecules. For this reason, pathologic tissu

$$
S \sim \frac{dM_{x,y}}{dt} \sim M_0 \left(1 - e^{-T_R/T_1}\right) e^{-T_E/T_2},
$$

dipole–dipole interactions within the rapidly tumbling ment (RARE) [23.34
water molecules. For this reason, pathologic tissue usu-
first introduction was
ally shows up as hyperintense (bright) on T_2 -weighted lacked att water molecules. For this reason, pathologic tissue usu-

illy shows up as hyperintense (bright) on T_2 -weighted

lacked attention. Seven years

images. The signal amplitude that later defines the
 $S \sim \frac{dM_{x,y}}{dt} \sim M_0$ ally shows up as hyperintense (bright) on T_2 -weighted lacked attention

images. The signal amplitude that later defines the were searching

brightness of the according pixel is given by
 $S \sim \frac{dM_{x,y}}{dt} \sim M_0 \left(1 - e^{-T_R/T$ ges. The signal amplitude that later defines the were searching for a fast T_2 -local

threess of the according pixel is given by
 $S \sim \frac{dM_{x,y}}{dt} \sim M_0 \left(1 - e^{-T_R/T_1}\right) e^{-T_E/T_2}$, localizer revealed images with impr

ree brightness of the according pixel is given by
 $S \sim \frac{dM_{x,y}}{dt} \sim M_0 \left(1 - e^{-T_R/T_1}\right) e^{-T_E/T_2}$, based on progress in technology

where M_0 represents the maximum possible longiu-

dinal nuclear magnetization M_z that can $S \sim \frac{GM_{X,Y}}{dt} \sim M_0 \left(1 - e^{-T_R/T_1}\right) e^{-T_E/T_2}$, localizer revealed images
surviving acronyms for the
dinal nuclear magnetization M_z that can be achieved echo (TSE).
within the given voxel. M_z is converted to transverse where M_0 represents the maximum possible longitu-
dinal nuclear magnetization M_z that can be achieved echo (TS
within the given voxel. M_z is converted to transverse The
nuclear magnetization M_{xy} using a 90° RF e

Resonance Imaging 23.3 MRI – Basic Principles and Applications 449
multiple RF refocusing pulses. On viewing such im-
ages, it becomes obvious that image contrast is only
slightly changing, especially for images acquired Resonance Imaging 23.3 MRI – Basic Principles and Applications 449

multiple RF refocusing pulses. On viewing such im-

ages, it becomes obvious that image contrast is only

slightly changing, especially for images acquir Resonance Imaging 23.3 MRI – Basic Principles and Applications 449

multiple RF refocusing pulses. On viewing such im-

ages, it becomes obvious that image contrast is only

slightly changing, especially for images acquir Resonance Imaging 23.3 MRI – Basic Principles and Applications

multiple RF refocusing pulses. On viewing such im-

ages, it becomes obvious that image contrast is only

slightly changing, especially for images acquired w Resonance Imaging 23.3 MRI – Basic Principles and Applications

multiple RF refocusing pulses. On viewing such im-

ages, it becomes obvious that image contrast is only

slightly changing, especially for images acquired w Resonance Imaging 23.3 MRI – Basic Principles and Applications 449

multiple RF refocusing pulses. On viewing such im-

ages, it becomes obvious that image contrast is only

slightly changing, especially for images acquir Resonance Imaging 23.3 MRI – Basic Principles and Applications

multiple RF refocusing pulses. On viewing such im-

ages, it becomes obvious that image contrast is only

slightly changing, especially for images acquired w parameter Imaging 23.3 MRI – Basic Principles and Applications

tiple RF refocusing pulses. On viewing such im-

s, it becomes obvious that image contrast is only

htly changing, especially for images acquired with

echoe Resonance Imaging 23.3 MRI – Basic Principles and Applications

multiple RF refocusing pulses. On viewing such im-

ages, it becomes obvious that image contrast is only

slightly changing, especially for images acquired w multiple RF refocusing pulses. On viewing such images, it becomes obvious that image contrast is only slightly changing, especially for images acquired with late echoes. This observation led to the idea of acquiring anoth multiple RF refocusing pulses. On viewing such im-
ages, it becomes obvious that image contrast is only
slightly changing, especially for images acquired with
late echoes. This observation led to the idea of acquiring
ano

localizer revealed images with impressive quality. The , also demonstrate long

that can be accompany that a sequence has be

the rapidly tumbling

that rapidly tumbling

ment (RARE) [23.34]. The im-

pathologic tissue usu-

first introduction was rather m

ight) on T_2 -weigh multiple RF refocusing pulses. On viewing such im-
ages, it becomes obvious that image contrast is only
slightly changing, especially for images acquired with
late echoes. This observation led to the idea of acquiring
ano ages, it becomes obvious that image contrast is only
slightly changing, especially for images acquired with
late echoes. This observation led to the idea of acquiring
another k-space line rather than producing another imslightly changing, especially for images acquired with
late echoes. This observation led to the idea of acquiring
another k-space line rather than producing another im-
age. This concept will potentially significantly red late echoes. This observation led to the idea of acquiring
another *k*-space line rather than producing another im-
age. This concept will potentially significantly reduce
the measurement time (Fig. 23.8).
Such a sequence another *k*-space line rather than producing another im-
age. This concept will potentially significantly reduce
the measurement time (Fig. 23.8).
Such a sequence has been introduced with the
acronym *rapid acquisition wi* age. This concept will potentially significantly reduce
the measurement time (Fig. 23.8).
Such a sequence has been introduced with the
acronym *rapid acquisition with relaxation enhance-*
ment (RARE) [23.34]. The image qu the measurement time (Fig. 23.8).

Such a sequence has been introduced with the

acronym *rapid acquisition with relaxation enhance-*

ment (RARE) [23.34]. The image quality at the time of

first introduction was rather m echo (TSE). mym *rapid acquisition with relaxation enhance*
 tu (RARE) [23.34]. The image quality at the time of

introduction was rather moderate, and the method

red attention. Seven years later, *Melki* and *Mulkern*

e searchin ment (RARE) [23.34]. The image quality at the time of
first introduction was rather moderate, and the method
lacked attention. Seven years later, *Melki* and *Mulkern*
were searching for a fast T_2 -localizer and *redisc* first introduction was rather moderate, and the method
lacked attention. Seven years later, *Melki* and *Mulkern*
were searching for a fast T_2 -localizer and *rediscov-*
ered the multi-echo spin-echo approach [23.35]. lacked attention. Seven years later, *Melki* and *Mulkern*
were searching for a fast T_2 -localizer and *rediscov-*
ered the multi-echo spin-echo approach [23.35]. Likely
based on progress in technology, implementation o

 $S \sim \frac{dM_{x,y}}{dt} \sim M_0 \left(1 - e^{-T_R/T_1}\right) e^{-T_E/T_2}$, based on progress in technology, in

where M_0 represents the maximum possible longitu-

diration and nuclear magnetization M_z that can be achieved

within the given voxe dinal nuclear magnetization M_z that can be achieved
within the given voxel. M_z is converted to transverse The advantage of this
nuclear magnetization M_{xy} using a 90° RF excitation duction in measurement t
pulse. M nuclear magnetization M_{xy} using a 90° RF excitation duction in measurement time
pulse. M_{xy} rotates with the Larmor frequency, inducing proportional to the number of
a signal in an adjacent receiver coil.
Introductio pulse. M_{xy} rotates with the Larmor frequency, inducing proportional to the number of echc
a signal in an adjacent receiver coil. Can called *echo train length* (ETL). A c
introduction of T_1 -shortening paramagnetic c a signal in an adjacent receiver coil. Called *echo train length* (ET

Diagnostic confidence has been increased with the be expected due to the fact

introduction of T_1 -shortening paramagnetic contrast a different *wei* were searching for a fast T_2 -localizer and *rediscov-*
ered the multi-echo spin-echo approach [23.35]. Likely
based on progress in technology, implementation of the
localizer revealed images with impressive quality. Th *ered* the multi-echo spin-echo approach [23.35]. Likely
based on progress in technology, implementation of the
localizer revealed images with impressive quality. The
surviving acronyms for the sequences that evolved ou based on progress in technology, implementation of the *localizer* revealed images with impressive quality. The surviving acronyms for the sequences that evolved out of this approach are *fast spin echo* (FSE), and *turbo localizer* revealed images with impressive quality. The
surviving acronyms for the sequences that evolved out
of this approach are *fast spin echo* (FSE), and *turbo spin*
echo (TSE).
The advantage of this method is a si surviving acronyms for the sequences that evolved out
of this approach are *fast spin echo* (FSE), and *turbo spin*
echo (TSE).
The advantage of this method is a significant re-
duction in measurement time, where the redu of this approach are *fast spin echo* (FSE), and *turbo spin*
echo (TSE).
The advantage of this method is a significant re-
duction in measurement time, where the reduction is
proportional to the number of echoes utilized *echo* (TSE).
The advantage of this method is a significant re-
duction in measurement time, where the reduction is
proportional to the number of echoes utilized, the so-
called *echo train length* (ETL). A disadvantage sh The advantage of this method is a significant re-
duction in measurement time, where the reduction is
proportional to the number of echoes utilized, the so-
called *echo train length* (ETL). A disadvantage should
be expec duction in measurement time, where the reduction is
proportional to the number of echoes utilized, the so-
called *echo train length* (ETL). A disadvantage should
be expected due to the fact that each Fourier line has
a di proportional to the number of echoes utilized, the so-
called *echo train length* (ETL). A disadvantage should
be expected due to the fact that each Fourier line has
a different *weighting*. Theoretically this could lead t Resonance Imaging | 23.3 $MRl = Basic Principles and Applications$
sages, it becomes obvious that image contrast is only
slightly changing, especially for images acquired with
slightly changing, expecting for images acquired with
and the t-space line en

Fig. 23.8 Sequence diagram
for a multi-echo spin-echo
sequence [turbo spin echo]
(TSE) fast spin echo (ESE)] Fig. 23.8 Sequence diagram
for a multi-echo spin-echo
sequence [turbo spin echo
(TSE), fast spin echo (FSE)] Fig. 23.8 Sequence diagram
for a multi-echo spin-echo
sequence [turbo spin echo
(TSE), fast spin echo (FSE)] Fig. 23.8 Sequence diagram
for a multi-echo spin-echo
sequence [turbo spin echo
(TSE), fast spin echo (FSE)]

Medical Imaging
mentioned potential disadvantages of the multi-echo tude by using a low-angle
approach. Today PD- and T_2 -weighted imaging are excitation pulse. The signal
solely done using FSE and TSE sequences.
exci **Medical Imaging**
 Applicance Specify Contains the median Contract Control
 Approach. Today PD- and T₂-weighted imaging are excitation pulse. The signal response

solely done using FSE and TSE sequences.
 APPENDIMA Medical Imaging
mentioned potential disadvantages of the multi-echo
approach. Today PD- and T_2 -weighted imaging are
solely done using FSE and TSE sequences.
23.3.4 The Gradient Echo Sequence Medical Imaging
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 EXECUTE: and T2-weighted imaging are excitation pulse.
 EXECUTE: excitation angle is
 23.3.4 The Gradient Echo Sequence

Medical Imaging

mentioned potential disadvantages of the multi-echo tude by using a low-angle excitation

approach. Today PD- and T₂-weighted imaging are excitation pulse. The signal response

solely done using FSE a **Medical Imaging**

mentioned potential disadvantages of the multi-echo tude by using a low-angle excitapproach. Today PD- and T_2 -weighted imaging are excitation pulse. The signal resolution solely done using FSE and TS **Medical Imaging**

mentioned potential disadvantages of the multi-echo tude by using a low-angle

approach. Today PD- and T₂-weighted imaging are excitation pulse. The signal

solely done using FSE and TSE sequences.
 mentioned potential disadvantages of the multi-echo tude by using a low-angle exapproach. Today PD- and T_2 -weighted imaging are excitation pulse. The signal resolution angle of the sum of such that is suggested the acr mentioned potential disadvantages of the multi-echo tude by using a low-angle excitation
approach. Today PD- and T_2 -weighted imaging are excitation pulse. The signal responss
solely done using FSE and TSE sequences. ex mentioned potential disadvantages of the multi-echo tude by using a low-angle excitation
approach. Today PD- and T₂-weighted imaging are excitation pulse. The signal responsely done using FSE and TSE sequences.

23.3.4 approach. Today PD- and T_2 -weighted imaging are excitation pulse. The signal isolely done using FSE and TSE sequences.
 23.3.4 The Gradient Echo Sequence $S \sim M_{x,y} = M_0 \frac{(1 - e^{-\beta})}{1 - \cos \alpha}$

Omitting the 180° RF refocu solely done using FSE and TSE sequences.
 23.3.4 The Gradient Echo Sequence
 $S \sim M_{x,y} = M_0 \frac{(1 - e^{-T_R/T_H})}{1 - \cos \alpha e^{-T_H}}$

Omitting the 180° RF refocusing pulse leads to the GREs are used whenever short regeneration of an *ec* **23.3.4 The Gradient Echo Sequence**
 $S \sim M_{x,y} = M_0 \frac{(1 - e^{-T_R/T_R})}{1 - \cos \alpha e^{-T_I}}$

Omitting the 180° RF refocusing pulse leads to the GREs are used whenever short regeneration of an *echo* by using bipolar gradient switch-

ing **23.3.4 The Gradient Echo Sequence**
 $S \sim M_{x,y} = M_0 \frac{1}{1 - \cos \alpha e^{-T_R}}$

Omitting the 180° RF refocusing pulse leads to the GREs are used whenever short regeneration of an *echo* by using bipolar gradient switch-sined. GREs a Omitting the 180° RF refocusing pulse leads to the GREs are used whenever short regeneration of an *echo* by using bipolar gradient switch-sired. GREs are used for fast imaging in conjunction with frequency encoding (Fig. Omitting the 180° RF refocusing pulse leads to the GREs are used whenever short repertion of an *echo* by using bipolar gradient switch-sired. GREs are used for fast imagin in conjunction with frequency encoding (Fig. 23. generation of an *echo* by using bipolar gradient switch-sired. GREs are used for fast imagin
ing in conjunction with frequency encoding (Fig. 23.6). T_2^* sensitivity is desired. As an exar
This suggested the acronym g ing in conjunction with frequency encoding (Fig. 23.6). T_2^* sensitivity is desired. As an exa This suggested the acronym gradient-echo sequence plied for imaging of the beating head (GRE). Different vendors use differ This suggested the acronym gradient-echo sequence plied for imaging of the beating h
(GRE). Different vendors use different acronyms for encoding is not limited to spatial
this generic approach, such as *fast low-angle sh* (GRE). Different vendors use different acronyms for encoding is not limited to spatial enchis generic approach, such as *fast low-angle shot* ond dimension within an imaging pl(FLASH) [23.37], *fast field echo* (FFE)-T1 o this generic approach, such as *fast low-angle shot* ond dimension (FLASH) [23.37], *fast field echo* (FFE)-T1 or *spoiled* be used to furt *gradient recalled acquired steady state* (SPGR). With- a volume. This out the us x_2^* . It is customary (FLASH) [23.37], *fast field echo* (FFE)-T1 or *spoiled* be used to further *partition* a sl
 gradient recalled acquired steady state (SPGR). With-

a volume. This approach is cal

out the use of a 180° RF refocusing pu **Part C** | Medical imaging

mentioned presentation disactumeges of the multi-schop make by using a low-stage exception instead of a SO°

and promod. Testig 15E and TSE sequences.
 $\frac{1}{2}$ and product in and TSE sequence

tude by using a low-angle excitation instead of a 90° excitation pulse. The signal response in relation to the excitation angle α in any given tissue follows tude by using a low-angle excitation instead of a 90°
excitation pulse. The signal response in relation to the
excitation angle α in any given tissue follows
 $S \sim M$ tude by using a low-angle excitation instead of a 90°
excitation pulse. The signal response in relation to the
excitation angle α in any given tissue follows
 $S \sim M_{x,y} = M_0 \frac{(1 - e^{-T_E/T_1})}{1 - \cos \alpha e^{-T_E/T_1}} e^{-T_E/T_2^*} \sin \alpha$.

$$
S \sim M_{x,y} = M_0 \frac{(1 - e^{-T_{\rm R}/T_1})}{1 - \cos \alpha e^{-T_{\rm R}/T_1}} e^{-T_{\rm E}/T_2^*} \sin \alpha.
$$

low-angle shot ond dimension within an in

E)-T1 or spoiled be used to further partition

(SPGR). With-

a volume. This approach is

soulse, the signal only disadvantage is that, for

molecular spin-

in the direction of tude by using a low-angle excitation instead of a 90°
excitation pulse. The signal response in relation to the
excitation angle α in any given tissue follows
 $S \sim M_{x,y} = M_0 \frac{(1 - e^{-T_R/T_1})}{1 - \cos \alpha e^{-T_R/T_1}} e^{-T_E/T_2^*} \sin \alpha$.
GR tude by using a low-angle excitation instead of a 90°
excitation pulse. The signal response in relation to the
excitation angle α in any given tissue follows
 $S \sim M_{x,y} = M_0 \frac{(1 - e^{-T_R/T_1})}{1 - \cos \alpha e^{-T_R/T_1}} e^{-T_E/T_2^*} \sin \alpha$.
GR T_2^* sensitivity is desired e by using a low-angle excitation instead of a 90°

itation pulse. The signal response in relation to the

itation angle α in any given tissue follows
 $S \sim M_{x,y} = M_0 \frac{(1 - e^{-T_R/T_1})}{1 - \cos \alpha e^{-T_R/T_1}} e^{-T_E/T_2^*} \sin \alpha$.

Es are us tude by using a low-angle excitation instead of a 90°
excitation pulse. The signal response in relation to the
excitation angle α in any given tissue follows
 $S \sim M_{x,y} = M_0 \frac{(1 - e^{-T_R/T_1})}{1 - \cos \alpha e^{-T_R/T_1}} e^{-T_E/T_2^*} \sin \alpha$.
GR tude by using a low-angle excitation instead of a 90°
excitation pulse. The signal response in relation to the
excitation angle α in any given tissue follows
 $S \sim M_{x,y} = M_0 \frac{(1 - e^{-T_R/T_1})}{1 - \cos \alpha e^{-T_R/T_1}} e^{-T_E/T_2^*} \sin \alpha$.
GR tude by using a low-angle excitation instead of a 90°
excitation pulse. The signal response in relation to the
excitation angle α in any given tissue follows
 $S \sim M_{x,y} = M_0 \frac{(1 - e^{-T_R/T_1})}{1 - \cos \alpha e^{-T_R/T_1}} e^{-T_E/T_2^*} \sin \alpha$.
GR excitation pulse. The signal response in relation to the
excitation angle α in any given tissue follows
 $S \sim M_{x,y} = M_0 \frac{(1 - e^{-T_R/T_1})}{1 - \cos \alpha e^{-T_R/T_1}} e^{-T_E/T_2^*} \sin \alpha$.
GREs are used whenever short repetition times are de-
si excitation angle α in any given tissue follows
 $S \sim M_{x,y} = M_0 \frac{(1 - e^{-T_R/T_1})}{1 - \cos \alpha e^{-T_R/T_1}} e^{-T_E/T_2^*} \sin \alpha$.

GREs are used whenever short repetition times are de-

sired. GREs are used for fast imaging and/or wherever
 $S \sim M_{x,y} = M_0 \frac{(1 - e^{-T_R/T_1})}{1 - \cos \alpha e^{-T_R/T_1}} e^{-T_E/T_2^*} \sin \alpha$.

GREs are used whenever short repetition times are de-

sired. GREs are used for fast imaging and/or wherever
 T_2^* sensitivity is desired. As an example, GREs $S \sim M_{x,y} = M_0 \frac{(1 - C)^2}{1 - \cos \alpha e^{-T_R/T_1}} e^{-T_E/T_2^*} \sin \alpha$.

GREs are used whenever short repetition times are de-

sired. GREs are used for fast imaging and/or wherever
 T_2^* sensitivity is desired. As an example, GREs are a $1 - \cos \alpha e^{-i\kappa/4}$
GREs are used whenever short repetition times are de-
sired. GREs are used for fast imaging and/or wherever
 T_2^* sensitivity is desired. As an example, GREs are ap-
plied for imaging of the beating he GREs are used whenever short repetition times are de-
sired. GREs are used for fast imaging and/or wherever
 T_2^* sensitivity is desired. As an example, GREs are ap-
plied for imaging of the beating heart (Fig. 23.9). P sired. GREs are used for fast imaging and/or wherever T_2^* sensitivity is desired. As an example, GREs are applied for imaging of the beating heart (Fig. 23.9). Phase encoding is not limited to spatial encoding of the T_2^* sensitivity is desired. As an example, GREs are ap-
plied for imaging of the beating heart (Fig. 23.9). Phase
encoding is not limited to spatial encoding of the sec-
ond dimension within an imaging plane, but can plied for imaging of the beating heart (Fig. 23.9). Phase
encoding is not limited to spatial encoding of the sec-
ond dimension within an imaging plane, but can also
be used to further *partition* a slice, in this case ca encoding is not limited to spatial encoding of the
ond dimension within an imaging plane, but can
be used to further *partition* a slice, in this case ca
a volume. This approach is called 3-D imaging.
only disadvantage is on within an imaging plane, but can also
urther *partition* a slice, in this case called
his approach is called 3-D imaging. The
tage is that, for every phase-encoding step
ion of slice selection, all phase-encoding
the im

The T_2^* sensitivity is very

Fig. 23.9 Representation of a singe
timeframe of an image of a beating
heart. Measurement time was 9.28 s
with temporal resolution of 43 ms Fig. 23.9 Representation of a singe
timeframe of an image of a beating
heart. Measurement time was 9.28 s
with temporal resolution of 43 ms
(four chamber view, myocardial in Fig. 23.9 Representation of a singe
timeframe of an image of a beating
heart. Measurement time was 9.28 s
with temporal resolution of 43 ms
(four-chamber view, myocardial in-
farction: courtesy of the PI A 306 Fig. 23.9 Representation of a singe
timeframe of an image of a beating
heart. Measurement time was 9.28 s
with temporal resolution of 43 ms
(four-chamber view, myocardial in-
farction; courtesy of the PLA 306
Hospital Reij Fig. 23.9 Representation of a singe
timeframe of an image of a beating
heart. Measurement time was 9.28 s
with temporal resolution of 43 ms
(four-chamber view, myocardial in-
farction; courtesy of the PLA 306
Hospital, Bei Fig. 23.9 Representation of a singe
timeframe of an image of a beating
heart. Measurement time was 9.28 s
with temporal resolution of 43 ms
(four-chamber view, myocardial in-
farction; courtesy of the PLA 306
Hospital, Bei Fig. 23.9 Representation of a singe
timeframe of an image of a beating
heart. Measurement time was 9.28 s
with temporal resolution of 43 ms
(four-chamber view, myocardial in-
farction; courtesy of the PLA 306
Hospital, Bei

23.3 MRI – Basic Principles and Applications

Fig. 23.10 Axial view of the intracra-

mial vasculature generated without

use of a contrast agent. The contrast

is solely based on unsaturated blood 23.3 MRI – Basic Principles and Applications
 Fig. 23.10 Axial view of the intracra-

nial vasculature generated without

use of a contrast agent. The contrast

is solely based on unsaturated blood

flowing into the ima 23.3 MRI – Basic Principles and Applications 451
 Fig. 23.10 Axial view of the intracra-

mial vasculature generated without

use of a contrast agent. The contrast

is solely based on unsaturated blood

flowing into the 23.3 MRI – Basic Principles and Applications

Fig. 23.10 Axial view of the intracra-

mial vasculature generated without

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is solely based on unsaturated blood

flowing into the imagi 23.3 MRI – Basic Principles and Applications

Fig. 23.10 Axial view of the intracra-

nial vasculature generated without

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flowing into the imagi 23.3 MRI – Basic Principles and Applications

Fig. 23.10 Axial view of the intracra-

rial vasculature generated without

use of a contrast agent. The contrast

is solely based on unsaturated blood

flowing into the imagi **Fig. 23.10** Axial view of the intracra-
 Fig. 23.10 Axial view of the intracra-

nial vasculature generated without

use of a contrast agent. The contrast

is solely based on unsaturated blood

flowing into the imaging 23.3 MRI – Basic Principles and Applications
 Fig. 23.10 Axial view of the intracra-

mial vasculature generated without

use of a contrast agent. The contrast

is solely based on unsaturated blood

flowing into the ima

in *susceptibility-weighted imaging* (SWI) [23.38]. The echo imaging, the 180° RF ref
change in oxygen concentration within the blood vas-
the faster component of the rotati
culature of active brain regions is correlated w in *susceptibility-weighted imaging* (SWI) [23.38]. The echo imaging, the 180° RF ret
change in oxygen concentration within the blood vas-
culature of active brain regions is correlated with magnetization behind the in *susceptibility-weighted imaging* (SWI) [23.38]. The echo imaging, the 180° RF reformance in oxygen concentration within the blood vas-
culature of active brain regions is correlated with magnetization behind the slowe in *susceptibility-weighted imaging* (SWI) [23.38]. The echo imaging, the 180° RF rechange in oxygen concentration within the blood vas-
culature of active brain regions is correlated with magnetization behind the slower
 in *susceptibility-weighted imaging* (SWI) [23.38]. The echo imaging, the 180° RF refocustion
change in oxygen concentration within the blood vas-
the faster component of the rotating
alteration of the magnetic susceptibi in *susceptibility-weighted imaging* (SWI) [23.38]. The echo imaging, the 180° RF refocustion
cchange in oxygen concentration within the blood vas-
culature of active brain regions is correlated with magnetization behind in susceptibility-weighted imaging (SWI) [23.38]. The echo imaging, the 180° RF ref
change in oxygen concentration within the blood vas-
the faster component of the rotation
culature of active brain regions is correlated using T_2^* -sensitive protocols From the sensitive protocols [23.41, 42]. One further case the net signal regions from the slow understand and water of the magnetic susceptibility and is used time the echo reaches its maxime oxygenation-dependent imagin culature of active brain regions is correlated with magnetization behind the slow
alteration of the magnetic susceptibility and is used time the echo reaches its maxi
in *blood oxygenation-dependent imaging* (BOLD) to cle alteration of the magnetic susceptibility and is used

in *blood oxygenation-dependent imaging* (BOLD) to

clear magnetizations are in pha

visualize active brain regions. As this allows the func-

image shift in the dire in *blood oxygenation-dependent imaging* (BOLD) to clear magnetizations are in phase visualize active brain regions. As this allows the func-
image shift in the direction of spin-
tion of the brain to be documented, it is visualize active brain regions. As this allows the func-
image shift in the direction
tion of the brain to be documented, it is also called remain. In GRE imaging th
functional MRI (fMRI) [23.39,40]. In conjunction with p tion of the brain to be documented, it is also called remain. In GRE imaging there is *functional MRI* (fMRI) [23.39, 40]. In conjunction with pulse, and depending on the echo intravenous injection of paramagnetic contras functional MRI (fMRI) [23.39, 40]. In conjunction with pulse, and depending on the ecl
intravenous injection of paramagnetic contrast agents, a situation where the transverse
the correlated changes in magnetic susceptibil intravenous injection of paramagnetic contrast agents, a situation where the transverse nu
the correlated changes in magnetic susceptibility as within lipids will be in *opposed-pl*
a marker for the passage of the contras the correlated changes in magnetic susceptibility as within lipids will be in *opposed* a marker for the passage of the contrast agent can be verse nuclear magnetization wit used to trace and document perfusion deficits w a marker for the passage of the contrast agent can be verse nuclear magnetization with
used to trace and document perfusion deficits when a later echo they will be *in-phase* is
using T_2^* -sensitive protocols [23.41, 4 used to trace and document perfusion deficits when a later echo they will be *in-ph*, using T_2^* -sensitive protocols [23.41, 42]. One further case the net signal will be *i* feature of gradient-echo sequences is worth using T_2^* -sensitive protocols [23.41, 42]. One further case the net signal will be zero feature of gradient-echo sequences is worth mention-
feature of gradient-echo sequences is worth mention-
ing. This feature is im

Example 180^o RF refocusing pulse places
the faster component of the rotating transverse nuclear
magnetization behind the slower component, and at the
clear magnetizations are in phase again and only the echo imaging, the 180° RF refocusing pulse places
the faster component of the rotating transverse nuclear
magnetization behind the slower component, and at the
time the echo reaches its maximum, both transverse nu-
clear m echo imaging, the 180° RF refocusing pulse places
the faster component of the rotating transverse nuclear
magnetization behind the slower component, and at the
time the echo reaches its maximum, both transverse nu-
clear m echo imaging, the 180° RF refocusing pulse places
the faster component of the rotating transverse nuclear
magnetization behind the slower component, and at the
time the echo reaches its maximum, both transverse nu-
clear m echo imaging, the 180° RF refocusing pulse places
the faster component of the rotating transverse nuclear
magnetization behind the slower component, and at the
time the echo reaches its maximum, both transverse nu-
clear m echo imaging, the 180° RF refocusing pulse places
the faster component of the rotating transverse nuclear
magnetization behind the slower component, and at the
time the echo reaches its maximum, both transverse nu-
clear m echo imaging, the 180° RF refocusing pulse places
the faster component of the rotating transverse nuclear
magnetization behind the slower component, and at the
time the echo reaches its maximum, both transverse nu-
clear m **Example 180°** RF refocusing pulse places
the faster component of the rotating transverse nuclear
magnetization behind the slower component, and at the
time the echo reaches its maximum, both transverse nu-
clear magnetiz echo imaging, the 180° RF refocusing pulse places
the faster component of the rotating transverse nuclear
magnetization behind the slower component, and at the
time the echo reaches its maximum, both transverse nu-
c echo imaging, the 180° RF refocusing pulse places
the faster component of the rotating transverse nuclear
magnetization behind the slower component, and at the
time the echo reaches its maximum, both transverse nu-
clear echo imaging, the 180° RF refocusing pulse places
the faster component of the rotating transverse nuclear
magnetization behind the slower component, and at the
time the echo reaches its maximum, both transverse nu-
clear m the faster component of the rotating transverse nuclear
magnetization behind the slower component, and at the
time the echo reaches its maximum, both transverse nu-
clear magnetizations are in phase again and only the
imag magnetization behind the slower component, and at the
time the echo reaches its maximum, both transverse nu-
clear magnetizations are in phase again and only the
image shift in the direction of spatial encoding will
remain time the echo reaches its maximum, both transverse nu-
clear magnetizations are in phase again and only the
image shift in the direction of spatial encoding will
remain. In GRE imaging there is no RF refocusing
pulse, and clear magnetizations are in phase again and only the
image shift in the direction of spatial encoding will
remain. In GRE imaging there is no RF refocusing
pulse, and depending on the echo time, there will be
a situation w image shift in the direction of spatial encoding will
remain. In GRE imaging there is no RF refocusing
pulse, and depending on the echo time, there will be
a situation where the transverse nuclear magnetization
within lipi remain. In GRE imaging there is no RF refocusing
pulse, and depending on the echo time, there will be
a situation where the transverse nuclear magnetization
within lipids will be in *opposed-phase* with the trans-
verse n pulse, and depending on the echo time, there a situation where the transverse nuclear magne within lipids will be in *opposed-phase* with th verse nuclear magnetization within water, when a later echo they will be *in-phas* within lipids will be in *opposed-phase* with the trans-
verse nuclear magnetization within water, whereas at
a later echo they will be *in-phase* again. In the *antiphase*
case the net signal will be zero for a voxel, wit a later echo they will be *in-phase* again. In the *antiphase*
case the net signal will be zero for a voxel, with fat
and water showing approximately identical transverse
nuclear magnetizations. The corresponding pixel wil Resonance Imaging $|z_{33}|$ MRI – Basic Principles and Applications $\frac{32}{16}$

Fig. 23.30. MAI was degreed to simulate the most im-

use of a contrast agent. The concepts of $\frac{32}{16}$

while acquiring the data lines-of

Part C | Medical Imaging

Part C | Medical Imaging

Part of the company of the method strike of the strike **Medical Imaging**
portant imaging sequences for current routine clinical magnetic field gradient amplit
imaging. In any case, all existing imaging sequences phasing, resulting in a signal v
the GRE group [23.441] with dif **Medical Imaging**
 interproference
 interpretent in any case, all existing imaging sequences
 interpretent in any case, all existing imaging sequences
 cometed in the SE or
 interpretent imaging to either the SE Medical Imaging
portant imaging sequences for current routine clinical magnetic field gradient amp
imaging. In any case, all existing imaging sequences phasing, resulting in a signa
can be characterized as belonging to Medical Imaging
portant imaging sequences for current routine clinical magnetic field gradient ampli
imaging. In any case, all existing imaging sequences phasing, resulting in a signal
can be characterized as belonging to Medical Imaging
portant imaging sequences for current routine clinical magnetic field gradient amplitude
imaging. In any case, all existing imaging sequences phasing, resulting in a signal void
can be characterized as bel **Medical Imaging**

portant imaging sequences for current routine clinical magnetic field gradient amplitud

imaging. In any case, all existing imaging sequences phasing, resulting in a signal voice

the GRE group [23.44], **Medical Imaging**

portant imaging sequences for current routine clinical magnetic field gradient ampli

imaging. In any case, all existing imaging sequences phasing, resulting in a signal

can be characterized as belongi **Medical Imaging**

portant imaging sequences for current routine clinical magnetic field gradient amplimaging. In any case, all existing imaging sequences phasing, resulting in a signal can be characterized as belonging t **Medical Imaging**

portant imaging sequences for current routine clinical magnetic field gradient amplitud

imaging. In any case, all existing imaging sequences phasing, resulting in a signal voicin

can be characterized portant imaging sequences for current routine clinical magnetic field gradient ampliting
imaging. In any case, all existing imaging sequences phasing, resulting in a signal v
can be characterized as belonging to either th portant imaging sequences for current routine clinical magnetic field gradient amplitude vimaging. In any case, all existing imaging sequences phasing, resulting in a signal void [1] can be characterized as belonging to e portant imaging sequences for current routine clinical magnetic field gradient amplitude v
imaging. In any case, all existing imaging sequences phasing, resulting in a signal void [2
can be characterized as belonging to e imaging. In any case, all existing imaging sequences phasing, resulting in
can be characterized as belonging to either the SE or on the signal can be
the GRE group [23.44], with different types of hybrids. value
Worth men can be characterized as belonging to either the SE or on the signal can be character
the GRE group [23.44], with different types of hybrids. value
Worth mentioning is the possibility of manipulating the
longitudinal or tr the GRE group [23.44], with different types of hybrids. value
Worth mentioning is the possibility of manipulating the
longitudinal or transverse nuclear magnetization prior
to or within an imaging sequence. A classic exam Worth mentioning is the possibility of manipulating the

longitudinal or transverse nuclear magnetization prior

to or within an imaging sequence. A classic example

is the inversion of the longitudinal nuclear magneti-
 longitudinal or transverse nuclear magnetization prior $b = \gamma^2 G_{\text{DW}}^2 \delta^2 \left(\Delta - \frac{1}{3} \right)$
to or within an imaging sequence. A classic example
is the inversion of the longitudinal nuclear magneti- with G_{DW} being t to or within an imaging sequence. A classic example
is the inversion of the longitudinal nuclear magneti-
zation prior to starting the imaging sequence. Such weighting magnetic field
inversion allows the signal of a tissu is the inversion of the longitudinal nuclear magneti-
vith G_{DW} being the amplitud
zation prior to starting the imaging sequence. Such weighting magnetic field gradie
inversion allows the signal of a tissue with a speci zation prior to starting the imaging sequence. Such weighting magnetic fiel
inversion allows the signal of a tissue with a specific duration of the magnetic T_1 relaxation time to be nulled. Tissue is only able to the t inversion allows the signal of a tissue with a specific duration of the magnetic field gradien T_1 relaxation time to be nulled. Tissue is only able to the temporal distance between the two emit a signal if longitudinal T₁ relaxation time to be nulled. Tissue is only able to the temporal distance between the twent
emit a signal if longitudinal nuclear magnetization is routine clinical imaging, b-values of
available at the time of excit emit a signal if longitudinal nuclear magnetization is routine clinical imaging, *b*-
available at the time of excitation. The RF excitation are customary.
pulse will convert the longitudinal nuclear magnetiza-
tion to tr available at the time of excitation. The RF excitation are customary.

pulse will convert the longitudinal nuclear magnetiza-

tion to transverse nuclear magnetization which rotates

with the Larmor frequency, inducing an pulse will convert the longitudinal nuclear magnetiza-

tion to transverse nuclear magnetization which rotates

with the Larmor frequency, inducing an MR signal in

a coil adjacent to the object to be imaged. Following wi tion to transverse nuclear magnetization which rotates

with the Larmor frequency, inducing an MR signal in

a coil adjacent to the object to be imaged. Following with *D* being the *apparent diffusion*

inversion, the in with the Larmor frequency, inducing an MR signal in
a coil adjacent to the object to be imaged. Following with *D* being the *apparent diffusio*
inversion, the inverted longitudinal nuclear magnetiza-
As the diffusion is a coil adjacent to the object to be imaged. Following with *D* being the *apparent diffusio* inversion, the inverted longitudinal nuclear magnetiza-
tion will approach the parallel alignment with the main (DTI) will enabl inversion, the inverted longitudinal nuclear magnetiza-

ition will approach the parallel alignment with the main (DTI) will enable the measureme

magnetic field within a time given by the T_1 relax-diffusional directio tion will approach the parallel alignment with the main (DTI) will enable the measurer magnetic field within a time given by the T_1 relax-
atiffusional direction [23.48] with
ation time of the tissue. There is a point magnetic field within a time given by the T_1 relax-
ation time of the tissue. There is a point in time at
which the longitudinal magnetization will be zero, oc-
curring at the point of transition between antiparallel
t In a time given by the T_1 relax-

affusional direction [23.48] with a

sue. There is a point in time at ical representation [23.49]. The p

all magnetization will be zero, oc-

direction of water molecules seen

of tra ation time of the tissue. There is a point in time at ical representation [23.49]. The premetion the longitudinal magnetization will be zero, occurring at the point of transition between antiparallel nerve sheets. The dif which the longitudinal magnetization will be zero, oc-

curring at the point of transition between antiparallel

nerve sheets. The diffusional directi

to parallel alignment. If the excitation pulse is placed

molecules i curring at the point of transition between antiparallel nerve sheets. The diffusional directo parallel alignment. If the excitation pulse is placed molecules indirectly allows the diffusation time will not be excited. Lip to parallel alignment. If the excitation pulse is placed molecules indirectly allows the dispiter at that point in time, that tissue with a specific T_1 tracts.

relaxation time will not be excited. Lipids have a relati at that point in time, that tissue with a specific T_1 tracts.

relaxation time will not be excited. Lipids have a rel-

atively short T_1 relaxation time. On selecting a short **23.3.6 MRI Spectroscopy**

time period b relaxation time will not be excited. Lipids have a rel-
atively short T_1 relaxation time. On selecting a short **23.3.6 MRI Spectro**:
time period between inversion and excitation pulse, e.g.,
150 ms, the signal from lip atively short T_1 relaxation time. On selecting a short **23.3.6 MRI Spectroscopy**
time period between inversion and excitation pulse, e.g.,
150 ms, the signal from lipids will be suppressed. The Different nuclei have di time period between inversion and excitation pulse, e.g.,
150 ms, the signal from lipids will be suppressed. The Different nuclei have different
time between inversion and excitation pulse is called and identical nuclei h 150 ms, the signal from lipids will be suppressed. The Different nuclei have different I
time between inversion and excitation pulse is called and identical nuclei have different
the inversion time T_1 . As a 150 ms inve time between inversion and excitation pulse is called and identical nuclei have diffe
the inversion time T_1 . As a 150 ms inversion time was depending on their electronic
considered short at the time, the acronym shortthe inversion time T_1 . As a 150 ms inversion time was depending on their electronic envirce considered short at the time, the acronym short-tau in-
nomenon is exploited in MR spect version recovery (STIR) was establish considered short at the time, the acronym short-tau in-

version recovery (STIR) was established [23.45]. At the

measure the levels of differ

other end of the scale of relaxation times is the rela-

tissues. The shift i follows a protocol is helpful to identify benign cystic lesions such as *N*-acetyl-
and to identify periventricular lesions that might oth-shifted by 2 ppm
erwise remain unnoticed in the presence of adjacent water. Fourier tra
br 33 MrC bested issolary

applied to make the correspondent interaction interaction interaction interaction in the presence of magnetic field gradients of magnetic field gradients of magnetic field gradients of magnetic fie

$$
M_z = M_0 \left(1 - 2e^{-T_1/T_1} + 2e^{-(T_R - T_E/2)/T_1} \right)
$$
tration on
enables
the

magnetic field gradient amplitude will lead to a de-
phasing, resulting in a signal void [23.47]. The effect
on the signal can be characterized by the so-called b-
value magnetic field gradient amplitude will lead to a dephasing, resulting in a signal void [23.47]. The effect on the signal can be characterized by the so-called *b*-value magnetic field gradient amplitude will lead to a de-
phasing, resulting in a signal void [23.47]. The effect
on the signal can be characterized by the so-called b-
value
 $b = v^2 G_{\text{row}}^2 \delta^2 \left(\Delta - \frac{\delta}{v} \right)$. magnetic field gradient amplitude will lead to a de-
phasing, resulting in a signal void [23.47]. The effect
on the signal can be characterized by the so-called *b*-
value
 $b = \gamma^2 G_{DW}^2 \delta^2 \left(\Delta - \frac{\delta}{3} \right)$,
with G_{DW} b

$$
b = \gamma^2 G_{\text{DW}}^2 \delta^2 \left(\Delta - \frac{\delta}{3} \right) \,,
$$

magnetic field gradient amplitude will lead to a de-
phasing, resulting in a signal void [23.47]. The effect
on the signal can be characterized by the so-called b-
value
 $b = \gamma^2 G_{\text{DW}}^2 \delta^2 \left(\Delta - \frac{\delta}{3} \right)$,
with $G_{\text{DW$ magnetic field gradient amplitude will lead to a de-
phasing, resulting in a signal void [23.47]. The effect
on the signal can be characterized by the so-called *b*-
value
 $b = \gamma^2 G_{DW}^2 \delta^2 \left(\Delta - \frac{\delta}{3} \right)$,
with G_{DW} b magnetic field gradient amplitude will lead to a de-
phasing, resulting in a signal void [23.47]. The effect
on the signal can be characterized by the so-called *b*-
value
 $b = \gamma^2 G_{DW}^2 \delta^2 \left(\Delta - \frac{\delta}{3} \right)$,
with G_{DW} b magnetic field gradient amplitude will lead to a de-
phasing, resulting in a signal void [23.47]. The effect
on the signal can be characterized by the so-called b-
value
 $b = \gamma^2 G_{DW}^2 \delta^2 \left(\Delta - \frac{\delta}{3} \right)$,
with G_{DW} bei phasing, resulting in a signal void [23.47]. Ton the signal can be characterized by the so-
value
 $b = \gamma^2 G_{DW}^2 \delta^2 \left(\Delta - \frac{\delta}{3} \right)$,
with G_{DW} being the amplitude of the on-
weighting magnetic field gradient, δ repr the signal can be characterized by the so-called b-

le
 $b = \gamma^2 G_{DW}^2 \delta^2 \left(\Delta - \frac{\delta}{3} \right)$,

1 Gpw being the amplitude of the diffusion-

ghting magnetic field gradient, δ represents the

ation of the magnetic field g $b = \gamma^2 G_{\text{DW}}^2 \delta^2 \left(\Delta - \frac{1}{3} \right)$,
with G_{DW} being the amplitude of the diffusion-
weighting magnetic field gradient, δ represents the
duration of the magnetic field gradient, and Δ indicates
the temporal d with G_{DW} being the amplitude of the diffusion-
weighting magnetic field gradient, δ represents the
duration of the magnetic field gradient, and Δ indicates
the temporal distance between the two gradient lobes. In

,

$$
S \sim e^{-bD} ,
$$

with G_{DW} being the amplitude of the diffusion-
weighting magnetic field gradient, δ represents the
duration of the magnetic field gradient, and Δ indicates
the temporal distance between the two gradient lobes. In weighting magnetic field gradient, δ represents the
duration of the magnetic field gradient, and Δ indicates
the temporal distance between the two gradient lobes. In
routine clinical imaging, *b*-values of up to 10 duration of the magnetic field gradient, and Δ indicates
the temporal distance between the two gradient lobes. In
routine clinical imaging, b-values of up to 1000 s/mm^2
are customary.
The signal is as follows
 $S \sim e$ the temporal distance between the two gradient lobes. In
routine clinical imaging, *b*-values of up to 1000 s/mm^2
are customary.
The signal is as follows
 $S \sim e^{-bD}$,
with *D* being the *apparent diffusion coefficient* routine clinical imaging, *b*-values of up to 1000 s/mm^2
are customary.
The signal is as follows
 $S \sim e^{-bD}$,
with *D* being the *apparent diffusion coefficient* (ADC).
As the diffusion is a tensor, *diffusion tensor i* are customary.

The signal is as follows
 $S \sim e^{-bD}$,

with *D* being the *apparent diffusion coefficient* (ADC).

As the diffusion is a tensor, *diffusion tensor imaging*

(DTI) will enable the measurement of the preferr tracts. with *D* being the *apparent diffusion coefficient* (ADC).
As the diffusion is a tensor, *diffusion tensor imaging*
(DTI) will enable the measurement of the preferred
diffusional direction [23.48] with an according graph-As the diffusion is a tensor, *diffusion tensor imaging*
(DTI) will enable the measurement of the preferred
diffusional direction [23.48] with an according graph-
ical representation [23.49]. The preferred diffusional
dir (DTI) will enable the measurement of the preferred
diffusional direction [23.48] with an according graph-
ical representation [23.49]. The preferred diffusional
direction of water molecules seems to be parallel to
nerve sh diffusional direction [23.48] with an according graphical representation [23.49]. The preferred diffusional
direction of water molecules seems to be parallel to
nerve sheets. The diffusional directivity of these water
mole

attenuated inversion recovery (FLAIR) [23.46]. Such nance frequencies of hydrogen nuc
a protocol is helpful to identify benign cystic lesions such as N-acetyl-aspartate (NAA) fo
and to identify periventricular lesions tha and to identify periventricular lesions that might oth-

erwise remain unnoticed in the presence of adjacent water. Fourier transformation f

bright signal from cerebrospinal fluid. The recovery of tion without a readout ical representation [23.49]. The preferred diffusional
direction of water molecules seems to be parallel to
nerve sheets. The diffusional directivity of these water
molecules indirectly allows the display of nerve fiber
tr direction of water molecules seems to be parallel to
nerve sheets. The diffusional directivity of these water
molecules indirectly allows the display of nerve fiber
tracts.
23.3.6 MRI Spectroscopy
Different nuclei have d nerve sheets. The diffusional directivity of these water
molecules indirectly allows the display of nerve fiber
tracts.
23.3.6 MRI Spectroscopy
Different nuclei have different Larmor frequencies,
and identical nuclei hav molecules indirectly allows the display of nerve fiber
tracts.
 23.3.6 MRI Spectroscopy

Different nuclei have different Larmor frequencies,

and identical nuclei have different Larmor frequencies

depending on their ele **23.3.6 MRI Spectroscopy**
 23.3.6 MRI Spectroscopy

Different nuclei have different Larmor frequencies,

and identical nuclei have different Larmor frequencies

depending on their electronic environment. This phe-

nome **23.3.6 MRI Spectroscopy**

Different nuclei have different Larmor frequencies,

and identical nuclei have different Larmor frequencies

depending on their electronic environment. This phe-

nomenon is exploited in MR spec **23.3.6 MRI Spectroscopy**

Different nuclei have different Larmor frequencies,

and identical nuclei have different Larmor frequencies

depending on their electronic environment. This phe-

nomenon is exploited in MR spec Different nuclei have different Larmor frequencies,
and identical nuclei have different Larmor frequencies
depending on their electronic environment. This phe-
nomenon is exploited in MR spectroscopy (MRS) to
measure the l Different nuclei have different Larmor frequencies,
and identical nuclei have different Larmor frequencies
depending on their electronic environment. This phe-
nomenon is exploited in MR spectroscopy (MRS) to
measure the l and identical nuclei have different Larmor frequencies
depending on their electronic environment. This phe-
nomenon is exploited in MR spectroscopy (MRS) to
measure the levels of different metabolites in body
tissues. The depending on their electronic environment. This phe-
nomenon is exploited in MR spectroscopy (MRS) to
measure the levels of different metabolites in body
tissues. The shift in Larmor frequency as a function
of the electron nomenon is exploited in MR spectroscopy (MRS) to
measure the levels of different metabolites in body
tissues. The shift in Larmor frequency as a function
of the electronic environment is usually described in
parts per mill measure the levels of different metabolites in body
tissues. The shift in Larmor frequency as a function
of the electronic environment is usually described in
parts per million (ppm) with respect to a reference
frequency (tissues. The shift in Larmor frequency as a function
of the electronic environment is usually described in
parts per million (ppm) with respect to a reference
frequency (usually free water); for example, the reso-
nance fr of the electronic environment is usually described in
parts per million (ppm) with respect to a reference
frequency (usually free water); for example, the reso-
nance frequencies of hydrogen nuclei in amino acids
such as parts per million (ppm) with respect to a reference
frequency (usually free water); for example, the reso-
nance frequencies of hydrogen nuclei in amino acids
such as *N*-acetyl-aspartate (NAA) found in neurons are
shifte frequency (usually free water); for example, the reso-
nance frequencies of hydrogen nuclei in amino acids
such as N -acetyl-aspartate (NAA) found in neurons are
shifted by 2 ppm relative to the spectral line of free
wat nance frequencies of hydrogen nuclei in amino acids
such as *N*-acetyl-aspartate (NAA) found in neurons are
shifted by 2 ppm relative to the spectral line of free
water. Fourier transformation following data acquisi-
tion

- Safety-Relevant Aspects

Fig. 23.11 (a) Pro-

ton MRI

spectroscopy

- Chapter Contains and Chapter Container ton MRI Fig. 25.11 (a) Pro-
ton MRI
spectroscopy
of a healthy
prostate. (b) Pro-
 $\begin{bmatrix} 0 & 0 \\ 0 & 0 \end{bmatrix}$ - Safety-Relevant Aspects

Fig. 23.11 (a) Pro-

ton MRI

spectroscopy

of a healthy

prostate. (b) Pro-

ton MRI

spectroscopy of prostate. (b) Proton MRI Safety-Relevant Aspects

Fig. 23.11 (a) Pro-

ton MRI

spectroscopy

of a healthy

prostate. (b) Pro-

ton MRI

spectroscopy of

a prostate carci-

noma (courtesy

of the Univer-- Safety-Relevant Aspects

Fig. 23.11 (a) Pro-

ton MRI

spectroscopy

of a healthy

prostate. (b) Pro-

ton MRI

spectroscopy of

a prostate carci-

noma (courtesy

of the Univer-

sity Hospital - Safety-Relevant Aspects

Fig. 23.11 (a) Pro-

ton MRI

spectroscopy

of a healthy

prostate. (b) Pro-

ton MRI

spectroscopy of

a prostate carci-

noma (courtesy

of the Univer-

sity Hospital

Mannheim) Fig. 23.11 (a) Pro-
ton MRI
spectroscopy
of a healthy
prostate. (b) Pro-
ton MRI
spectroscopy of
a prostate carci-
noma (courtesy
of the Univer-
sity Hospital
Mannheim) Fig. 23.11 (a) Pro-
ton MRI
spectroscopy
of a healthy
prostate. (b) Pro-
ton MRI
spectroscopy of
a prostate carci-
noma (courtesy
of the Univer-
sity Hospital
Mannheim) Mannheim)

Besides proton spectroscopy, phosphorus spec-

produced by prostate tissue. Pro

troscopy should be mentioned. In this case, phosphorus sume citrate, and the intracellu

is utilized as the signal-emitting nucleus. Will be troscopy should be mentioned. In this case, phosphorus sume citrate, and the intracellul
is utilized as the signal-emitting nucleus. will be lowered. Choline is party
whereas phosphorus spectroscopy is utilized to study w Proton spectroscopy is mainly applied to the brain, brane and in malignant prostate le
whereas phosphorus spectroscopy is utilized to study will cell proliferation, the cholin
muscles, as phosphorus allows detection of ene whereas phosphorus spectroscopy is utilized to study will cell proliferation, the cholimuscles, as phosphorus allows detection of energy (Fig. 23.11).

23.4 MRI – Safety-Relevant Aspects

For the time period 1995–2005, the muscles, as phosphorus allows detection of energy (Fig.23.11).
 23.4 MRI – Safety-Relevant Aspects

For the time period 1995–2005, the Food and Drug Ad-
 EVALUAT COMENT COMENT COMENT COMENT COMENT COMENT COMENT CONS
 23.4 MRI – Safety-Relevant Aspects
For the time period 1995–2005, the Food and Drug Administration (FDA) database shows 389 entries where
humans have been harmed in conjunction with MRI.
Ten percent of these accidents w **23.4 MRI – Safety-Relevant Aspects**

For the time period 1995–2005, the Food and Drug Ad-
 EXECUTE COM FOLA) database shows 389 entries where

Internation (FDA) database shows 389 entries where

The percent of these ac For the time period 1995–2005, the Food and Drug Ad-

ministration (FDA) database shows 389 entries where

the magnetic of these accidents were related to ferro-

magnetic objects attracted by the strong magnetic field

S For the time period 1995–2005, the Food and Drug Advantual Cold gases in coministration (FDA) database shows 389 entries where

the ministration (FDA) database shows 389 entries where

the patient of these accidents were ministration (FDA) database shows 389 entries where

humans have been harmed in conjunction with MRI.

Ten percent of these accidents were related to ferro-

tion with Gd-contain

magnetic objects attracted by the strong humans have been harmed in conjunction with MRI.

• Nephrogenic systemic fibrosis (Ten percent of these accidents were related to ferro-

ion with Gd-containing contrast

magnetic objects attracted by the strong magnetic percent of these accidents were related to ferro-
tion with Gd-containing completic objects attracted by the strong magnetic field.
enty percentage were related to burns caused by RF 23.4.1 Attraction and Tourarized as fo Seventy percentage were related to burns caused by RF 23.4.1 Attraction and To
interactions. The most important safety issues can be
summarized as follows:

Attractive forces by the strong magnetic field
significant torq

-
-
-
-
- amplitudes
- ractions. The most important safety issues can be

marized as follows:

Attractive forces by the strong magnetic field

Significant torques within the storng magnetic field

RF interaction with the patient's body

RF inte Attractive forces by the strong magnetic field

Attractive forces by the strong magnetic field

RE interaction with the patient's body

RE interaction with active or passive implants

Peripheral nerve stimulation (PNS) ca

Propendicular to each other. The final signal will metabolism. In confirming the dependent of the voxel represented by the intersection carcinoma, proton spectroscopy is the direct of the control of the citrate peak and an are perpendicular to each other. The final signal will metabolism. In confirming the diome out of the voxel represented by the intersection carcinoma, proton spectroscopy of the three orthogonal slices. The currently achie are perpendicular to each other. The final signal will metabolism. In confirming the come out of the twoxel represented by the intersection carcinoma, proton spectroscopy of the three orthogonal slices. The currently achie extra a non-
 performance of the compared with a normal prostate

peak as compared with a normal prostate. Citrate is

produced by prostate tissue. Prostate carcinomas con-

will be lowered. Choline is north of the cell produced by prostate tissue. Prostate careful and the intracellular content of circle and a prostate carcinomas proton spectroscopy demonstrates a lower-
produced by prostate tissue. Prostate carcinomas con-
sume citrate, sumertabolism. In confirming the diagnosis of a prostate
carcinoma, proton spectroscopy demonstrates a lower-
ing of the citrate peak and an increase of the choline
peak as compared with a normal prostate. Citrate is
produ will be lowered. The continuing the diagnosis of a prostate
carcinoma, proton spectroscopy demonstrates a lower-
ing of the citrate peak and an increase of the choline
peak as compared with a normal prostate. Citrate is
pr **Example 18 The Example and in malignant prostate**
carcinoma, proton spectroscopy demonstrates a lower-
ing of the citrate peak and an increase of the choline
peak as compared with a normal prostate. Citrate is
produced by metabolism. In confirming the diagnosis of a prostate carcinoma, proton spectroscopy demonstrates a lower-
ing of the citrate peak and an increase of the choline
peak as compared with a normal prostate. Citrate is
produced (Fig. 23.11). come out of the voxel represented by the intersection carcinoma, proton spectros
of the three orthogonal slices. The currently achievable ing of the citrate peak and
voxel size is about 2–8 ml (for proton spectroscopy). b For the time period 1995–2005, the Food and Drug Ad-

The time the time period of the time and the time cellular

is utilized as the signal-emitting nucleus. Will be lowered. Choline is part

Froton spectroscopy is mainly in peak as compared with a normal prostate. Citrate is
produced by prostate tissue. Prostate carcinomas con-
sume citrate, and the intracellular content of citrate
will be lowered. Choline is part of the cell mem-
brane an sume citrate, and the intracellular content of citrate

will be lowered. Choline is part of the cell mem-

brane and in malignant prostate lesions, in conjunction

will cell proliferation, the choline level is increased
 the lowered. Choline is part of the cell mem-

in and in malignant prostate lesions, in conjunction

cell proliferation, the choline level is increased

[.23.11].

Cold gases in case of loss of superconductivity

(quenchin will cell proliferation, the choline level is increased

(Fig. 23.11).

• Cold gases in case of loss of superconductivity

(quenching)

• Nephrogenic systemic fibrosis (NSF), in conjunc-

tion with Gd-containing contrast a

- (quenching)
- 11).

gases in case of loss of superconductivity

orgenic systemic fibrosis (NSF), in conjunc-

with Gd-containing contrast agents.
 **Attraction and Torque

Due to Strong Magnetic Fields**

e forces on ferromagnetic objects

• Cold gases in case of loss of superconductivity

(quenching)

• Nephrogenic systemic fibrosis (NSF), in conjunc-

tion with Gd-containing contrast agents.
 23.4.1 Attraction and Torque
 Due to Strong Magnetic Fields • Cold gases in case of loss of superconductivity

(quenching)

• Nephrogenic systemic fibrosis (NSF), in conjunc-
 23.4.1 Attraction and Torque
 Due to Strong Magnetic Fields

Attractive forces on ferromagnetic objec • Nephrogenic systemic fibrosis (NSF), in conjunc-
tion with Gd-containing contrast agents.
 23.4.1 Attraction and Torque
 Due to Strong Magnetic Fields

Attractive forces on ferromagnetic objects are a conse-

quence

$$
F=\nabla U.
$$

$$
U=\frac{1}{2}M\mathbf{B}_0\,;\quad \, \mathbf{M}=\frac{\chi}{\mu_0}\,V\mathbf{B}_0\,.
$$

(quenching)

• Nephrogenic systemic fibrosis (NSF), in conjunc-
 EXECUTE: The potential energy is the product of the magnetic mo-
 EXECUTE: The potential energy is the product of the magnetic mo-

FEV U.

The potentia **23.4.1 Attraction and Torque**
 Due to Strong Magnetic Fields

Attractive forces on ferromagnetic objects are a consequence of a change in energy as a function of location
 $F = \nabla U$.

The potential energy is the produc **Due to Strong Magnetic Fields**
Attractive forces on ferromagnetic objects are a consequence of a change in energy as a function of location
 $F = \nabla U$.
The potential energy is the product of the magnetic mo-
ment and the

Medical Imaging

location B_0 . Simplified to the view of a single dimen-

sion, the attractive force on a ferromagnetic object is

proportional to the magnetic field gradient at the current will refise to start a mea **Medical Imaging**
location B_0 . Simplified to the view of a single dimen-
sion, the attractive force on a ferromagnetic object is runner. All vendors have to ensure the
proportional to the magnetic field gradient at the **Medical Imaging**

location B_0 . Simplified to the view of a single dimen-

sion, the attractive force on a ferromagnetic object is

proportional to the magnetic field gradient at the current will refuse to start a mea Medical Imaging

location B_0 . Simplified to the view of a single dimen-

sion, the attractive force on a ferromagnetic object is

proportional to the magnetic field gradient at the current will refuse to start a measur

$$
F_z = \frac{\chi V}{\mu_o} \mathbf{B}_0 \frac{\partial \mathbf{B}_0}{\partial z} \ .
$$

Free transmissions are all the view of a single dimensional to the view of a single dimension and this corresponds to the sion, the attractive force on a ferromagnetic object is trunner. All vendors have proportional to t force, interaction of the view of a single dimension, the attractive force on a ferromagnetic object is unner. All vendors have to ensproportional to the magnetic field gradient at the current will refuse to start a measu location B_0 . Simplified to the view of a single dimen-
sion, the attractive force on a ferromagnetic object is runner. All vendors have to ens
proportional to the magnetic field gradient at the current will refuse to s incoming a summer and the magnetic field gradient and the community of the may consider the may dividend at the community of the may dividend at the community of the maximum pulling force for a 1.5 T system is approximate sometic in the magnetic field gradient at the current will refuse to start a measurement
location and the magnetic field gradient at the current will refuse to start a measurement
location and the magnetic field strength proportional of the magnetic field strength at that location
and the magnetic field strength at that location
at the exposed to an energy level bey
docation and the magnetic field strength at that location
of patients wit net is zero. A ferromagnetic geometrically asymmetric
 $F_z = \frac{\chi V}{\mu_o} B_0 \frac{\partial B_0}{\partial z}$.

For a pair of scissors with weight of 0.41b, the hor-

ition of patients with perman

about 1.5% of these cases.

about 1.5% of these $F_z = \frac{\chi V}{\mu_o} B_0 \frac{\partial B_0}{\partial z}$.

For a pair of scissors with weight of 0.41b, the hor-

Studies have reported transient, sor

izontal maximum pulling force for a 1.5T system is

about 1.5% of these cases. If the particula For a pair of scissors with weight of 0.41b, the horizontal maximum pulling force for a 1.5 T system is
izontal maximum pulling force for a 1.5 T system is
approximately 20 times higher than the gravitational
obset to a w For a pair of scissors with weight of 0.41b, the hor-

intation, cutaneous swelling or h

intation, cutaneous swelling or h

about 1.5% of these cases. If the p

about 1.5% of these cases. If the p

about 1.5% of these ca izontal maximum pulling force for a 1.5T system is
about 1.5% of these cases. If the pati-
approximately 20 times higher than the gravitational
force, comparable to a weight of 8 lb. The critical aspect
painful sensations approximately 20 times higher than the gravitational point is the MR measurem
force, comparable to a weight of 8 lb. The critical aspect
force is experienced in the vicinity of the force, which is rapidly chang-
ing as a force, comparable to a weight of 8 lb. The critical aspect

is the discontinuity of the force, which is rapidly chang-

ing as a function of location. The maximum attractive

force is experienced in the vicinity of the pat ing as a function of location. The maximum attractive
force is experienced in the vicinity of the patient bore,
whereas the attractive force at the isocenter of the mag-
net is zero. A ferromagnetic geometrically asymmetri Experienced in the vicinity of the patient bore,

the attractive force at the isocenter of the mag-

o. A ferromagnetic geometrically asymmetric

Il have a strong tendency to align its long axis

of the direction of the m net is zero. A ferromagnetic geometrically asymmetric

object will have a strong tendency to align its long axis

parallel to the direction of the magnetic field. Ferromagnetic field at

parallel to the direction of the ma

23.4.2 RF Interaction
 23.4.2 RF Interactio 23.4.2 RF Interaction
 Examplement to the case for older pacemakers, older
 ITEL proportional to the Patient's Body
 ITEL proportional to the electromagnetic radiation
 ITEL proportional to the electromagnetic ra 23.4.2 RF Interaction
 and older anexy of the power spectrum of the electromagnetic radiation
 and older anexy melips. Orthopee
 and older anexy melips. Orthopee
 and on the magnetic component B1 of the electrom **With the Patient's Body**

general nonferromagnetic and do not

general nonferromagnetic and do not

used in MRI is too weak to lead to ionization, molecu-

lar destruction, or general nonferromagnetic and do not

lar des The power spectrum of the electromagnetic radiation

used in MRI is too weak to lead to ionization, moleculary hard

lare destruction, or generation of free-radical molecules.

The applied energy will only lead to warming The power spectrum of the electromagnetic radiation
used in MRI is too weak to lead to ionization, molecules.
lar destruction, or generation of free-radical molecules.
The applied energy will only lead to warming of the
p conductivity ρ portional to the square of the resonance frequency of

the system ω_0 (which is proportional to the magnetic

field strength used) and proportional to the magnetic

field strength used) and proportional to the square o the system ω_0 (which is proportional to the magnetic products. There are already a num
field strength used) and proportional to the square of dealing with the MR compatibility of
the magnetic component B_1 of the el field strength used) and proportional to the square of

the magnetic component B_1 of the electromagnetic ra-

diation used. The power W absorbed by the patient is plants, the concern is not lim

proportional to the fif

$$
W \approx \frac{\omega_0^2 B_1^2 b^5}{\rho} \ .
$$

the magnetic component B_1 of the electromagnetic ra-

diation used. The power W absorbed by the patient is

plants, the concern is not limited to

proportional to the fifth power of the circumference b

field. The coup diation used. The power W absorbed by the patient is
proportional to the fifth power of the circumference b
field. The coupling of the applie
of the patient and inversely proportional to the internal
conductivity ρ
con proportional to the fifth power of the circumference b

of the patient and inversely proportional to the internal

conductivity ρ
 $W \approx \frac{\omega_0^2 B_1^2 b^5}{\rho}$.
 $W \approx \frac{\omega_0^2 B_1^2 b^5}{\rho}$.
 $W \approx \frac{v_0^2 B_1^2 b^5}{\rho}$.
 $W \$ order the patient and inversely proportional to the internal

conductivity ρ
 $W \approx \frac{\omega_0^2 B_1^2 b^5}{\rho}$.
 $W \approx \frac{\omega_0^2 B_1^2 b^5}{\rho}$.

Acceptable levels of power deposition into a patient are

documented in the internat c.g., pacemaker leads, remains a rice of the duration $W \approx \frac{\omega_0^2 B_1^2 b^5}{\rho}$.
 $W \approx \frac{\omega_0^2 B_1^2 b^5}{\rho}$.

Acceptable levels of power deposition into a patient are

documented in the international guideline on safety r $W \approx \frac{\omega_0^2 B_1^2 b^5}{\rho}$.

Acceptable levels of power deposition into a patient are

documented in the international guideline on safety re-

documented in the international guideline on safety re-

quirements in MR (IEC $W \approx \frac{\omega_0 P_1 D}{\rho}$.

Acceptable levels of power deposition into a patient are

documented in the international guideline on safety re-

quirements in MR (IEC 60601-2-33). The metabolic

rate of the average patient is abo

and this corresponds to the metabolic rate of a marathon
runner. All vendors have to ensure that their systems
will refuse to start a measurement where the patient will
be exposed to an energy level beyond these quidelines and this corresponds to the metabolic rate of a marathon
runner. All vendors have to ensure that their systems
will refuse to start a measurement where the patient will
be exposed to an energy level beyond these guidelines and this corresponds to the metabolic rate of a marathon
runner. All vendors have to ensure that their systems
will refuse to start a measurement where the patient will
be exposed to an energy level beyond these guidelines and this corresponds to the metabolic rate of a marathon
runner. All vendors have to ensure that their systems
will refuse to start a measurement where the patient will
be exposed to an energy level beyond these guidelines this corresponds to the metabolic rate of a marathon
ner. All vendors have to ensure that their systems
refuse to start a measurement where the patient will
xposed to an energy level beyond these guidelines.
Another known

Nedical Imaging

location B_0 . Simplified to the view of a single dimen-

sion, the attractive force on a ferromagnetic object is runner. All vendors have to ensure

proportional to the magnetic field gradient at the **Medical Imaging**

location B_0 . Simplified to the view of a single dimen-

sion, the attractive force on a ferromagnetic object is runner. All vendors have to ensure tropportional to the magnetic field gradient at the and this corresponds to the metabolic rate of a marathon

runner. All vendors have to ensure that their systems

will refuse to start a measurement where the patient will

be exposed to an energy level beyond these guideli and this corresponds to the metabolic rate of a marathon

runner. All vendors have to ensure that their systems

will refuse to start a measurement where the patient will

be exposed to an energy level beyond these guideli and this corresponds to the metabolic rate of a marathon
runner. All vendors have to ensure that their systems
will refuse to start a measurement where the patient will
be exposed to an energy level beyond these guidelines and this corresponds to the metabolic rate of a marathon

runner. All vendors have to ensure that their systems

will refuse to start a measurement where the patient will

be exposed to an energy level beyond these guideli and this corresponds to the metabolic rate of a marathon

runner. All vendors have to ensure that their systems

will refuse to start a measurement where the patient will

be exposed to an energy level beyond these guideli and this corresponds to the metabolic rate of a marathon

runner. All vendors have to ensure that their systems

will refuse to start a measurement where the patient will

be exposed to an energy level beyond these guideli and this corresponds to the metabolic rate of a marathon
runner. All vendors have to ensure that their systems
will refuse to start a measurement where the patient will
be exposed to an energy level beyond these guidelines runner. All vendors have to ensure that their systems
will refuse to start a measurement where the patient will
be exposed to an energy level beyond these guidelines.
Another known potential complication is examina-
tion o will refuse to start a measurement where the patient
be exposed to an energy level beyond these guideli
Another known potential complication is exant
tion of patients with permanent cosmetics and tatt
Studies have reported Another known potential complication is examina-
tion of patients with permanent cosmetics and tattoos.
Studies have reported transient, sometimes painful skin
irritation, cutaneous swelling or heating sensations, in
about atients with permanent cosmetics and tattoos.
ave reported transient, sometimes painful skin
cutaneous swelling or heating sensations, in
% of these cases. If the patient communicates
ensations, the MR measurement can and irritation, cutaneous swelling or heating sensations, in
about 1.5% of these cases. If the patient communicates
painful sensations, the MR measurement can and will be
aborted by the technologist running the system. There
a about 1.5% of these cases. If the patient communicates
painful sensations, the MR measurement can and will be
aborted by the technologist running the system. There
are no reports that any of this damage is of perma-
nent n

object will have a strong tendency to align its long axis

parallel to the direction of the magnetic field. Ferromag-

netic scissors will always fly into the magnet with their

paral, all active and passive Impli-

parall parallel to the direction of the magnetic field. Ferromagnetic in a pointed tip entering first. The torque is proportional to the cancer of the square of the magnetic field strength and is at its dislodge ferromagnetic imp netic scissors will always fly into the magnet with their

pointed tip entering first. The torque is proportional to

the square of the magnetic field strength and is at its

maximum at the isocenter of the magnet.
 23.4. pointed tip entering first. The torque is proportional to
the square of the magnetic field strength and is at its
dislodge ferromagnetic implants, in aximum at the isocenter of the magnet.
23.4.2 RF Interaction
23.4.2 Example 1.1 Alternation to the square of the magnetic field strength and is at its dislodge ferromagnetic implants, if
maximum at the isocenter of the magnet.
 23.4.2 RF Interaction
 23.4.2 RF Interaction
 23.4.2 R painful sensations, the MR measurement can and will be
aborted by the technologist running the system. There
are no reports that any of this damage is of perma-
nent nature. Swelling, reddening or blistering have so
far be aborted by the technologist running the system. There
are no reports that any of this damage is of perma-
nent nature. Swelling, reddening or blistering have so
far been temporary.
23.4.3 Interaction with Active
and Pas are no reports that any of this damage is of perma-
nent nature. Swelling, reddening or blistering have so
far been temporary.
23.4.3 Interaction with Active
and Passive Implants
In general, all active and passive impl **and Passive Implants**
 And Solution and Passive Implants
 And Passive Implants

In general, all active and passive implants are of con-

cern [23.50]. The static magnetic field strength will

dislodge ferromagnetic **23.4.3 Interaction with Active**
 23.4.3 Interaction with Active
 and Passive Implants

In general, all active and passive implants are of con-

cern [23.50]. The static magnetic field strength will

dislodge ferromag **23.4.3 Interaction with Active**
 and Passive Implants

In general, all active and passive implants are of con-

cern [23.50]. The static magnetic field strength will

dislodge ferromagnetic implants, if the torque and a **23.4.3 Interaction with Active**
 and Passive Implants

In general, all active and passive implants are of con-

cern [23.50]. The static magnetic field strength will

dislodge ferromagnetic implants, if the torque and **and Passive Implants**
In general, all active and passive implants are of con-
cern [23.50]. The static magnetic field strength will
dislodge ferromagnetic implants, if the torque and at-
traction exceed the holding force. In general, all active and passive implants are of con-
cern [23.50]. The static magnetic field strength will
dislodge ferromagnetic implants, if the torque and at-
traction exceed the holding force. This is reported to be In general, all active and passive implants are of con-
cern [23.50]. The static magnetic field strength will
dislodge ferromagnetic implants, if the torque and at-
traction exceed the holding force. This is reported to be cern [23.50]. The static magnetic field strength will
dislodge ferromagnetic implants, if the torque and at-
traction exceed the holding force. This is reported to be
the case for older pacemakers, older cochlear implants, dislodge ferromagnetic implants, if the torque and at-
traction exceed the holding force. This is reported to be
the case for older pacemakers, older cochlear implants,
and older aneurysm clips. Orthopedic implants are in
 traction exceed the holding force. This is reported to be
the case for older pacemakers, older cochlear implants,
and older aneurysm clips. Orthopedic implants are in
general nonferromagnetic and do not present a con-
trai the case for older pacemakers, older cochlear implants,
and older aneurysm clips. Orthopedic implants are in
general nonferromagnetic and do not present a con-
traindication to MRI study. The static magnetic field
is also and older aneurysm clips. Orthopedic implants are in
general nonferromagnetic and do not present a con-
traindication to MRI study. The static magnetic field
is also potentially harmful to the function of active im-
plants general nonferromagnetic and do not present a con-
traindication to MRI study. The static magnetic field
is also potentially harmful to the function of active im-
plants such as older pacemakers. The industry involved
in d traindication to MRI study. The static magnetic field
is also potentially harmful to the function of active im-
plants such as older pacemakers. The industry involved
in design and production of implants is, of course, awa is also potentially harmful to the function of active im-
plants such as older pacemakers. The industry involved
in design and production of implants is, of course, aware
of the importance of MR for today's diagnostic ra-
 plants such as older pacemakers. The industry involved
in design and production of implants is, of course, aware
of the importance of MR for today's diagnostic ra-
diology, and is anxious to introduce MR-compatible
product of the importance of MR for today's diagnostic radiology, and is anxious to introduce MR-compatible products. There are already a number of publications dealing with the MR compatibility of some pacemaker designs and neuro and is anxious to introduce MR-compatible
There are already a number of publications
vith the MR compatibility of some pacemaker
nd neurostimulators. Especially for active im-
e concern is not limited to the static magneti Find with the MR compatibility of some pacemaker
Img with the MR compatibility of some pacemaker
gns and neurostimulators. Especially for active im-
tts, the concern is not limited to the static magnetic
I. The coupling of designs and neurostimulators. Especially for active im-
plants, the concern is not limited to the static magnetic
field. The coupling of the applied RF should not be un-
derestimated. The potential coupling of the RF with plants, the concern is not limited to the static magnetic
field. The coupling of the applied RF should not be un-
derestimated. The potential coupling of the RF with,
e.g., pacemaker leads, remains a risk, even if this is sphere the entropy is a the conduction of the spin of the orientation of the orientation of the orientation α conduction α conduction α conduction α conduction α conduction α conduction α conduction

$$
\oint_{\partial S} E \, \mathrm{d}l = -\frac{\partial}{\partial t} \iint_{S} B \, \mathrm{d}A \; .
$$

EXERCISE THE INTEREST IN Magnetic Field Gradients

Fipheral Nerve Stimulation (PNS)

w of induction indicates that a temporal change

magnetic field strength will induce a voltage
 $E dl = -\frac{\partial}{\partial t} \iint_S B dA$.

Iman body can Netive sumulation (PNS)

inction indicates that a temporal change

field strength will induce a voltage
 $\frac{\partial}{\partial t} \iint_S B \, dA$.

dy can present conductive loops, de-

induction. Depending on the orientation **23.4.4 Interaction Based on Changes**
 23.4.4 Interaction Based on Changes
 in Magnetic Field Gradients
 Peripheral Nerve Stimulation (PNS)

The law of induction indicates that a temporal change

in the magnetic fie Magnetic Resonance Imaging 23.4

of the switched magnetic field gradients, the ampli- in the now decaying magnetic fit

tudes and switching times of today's gradient systems to the liquid-helium bath, result

the human hod Magnetic Resonance Imaging 23

of the switched magnetic field gradients, the ampli- in the now decaying magnetic

tudes and switching times of today's gradient systems to the liquid-helium bath, resu

are capable of induci Magnetic Resonance Imaging 23.4

of the switched magnetic field gradients, the ampli-

in the now decaying magnetic fit

tudes and switching times of today's gradient systems to the liquid-helium bath, result

are capable Magnetic Resonance Imaging 23.4

of the switched magnetic field gradients, the ampli-

in the now decaying magnetic field

tudes and switching times of today's gradient systems to the liquid-helium bath, result

in are cap Magnetic Resonance Imaging 23.4 M

ally applied to estimate the muscle contractions and switching times of today's gradient systems to the liquid-helium bath, resulting

are capable of inducing voltages and currents within Magnetic Resonance Imaging 23.

of the switched magnetic field gradients, the ampli- in the now decaying magnetic fit

tudes and switching times of today's gradient systems to the liquid-helium bath, result

are capable o Magnetic Resonance Imaging 23.4 MR

of the switched magnetic field gradients, the ampli- in the now decaying magnetic field v

tudes and switching times of today's gradient systems to the liquid-helium bath, resulting

are Magnetic Resonan

of the switched magnetic field gradients, the ampli-

in the now deca;

tudes and switching times of today's gradient systems

to the liquid-hel

are capable of inducing voltages and currents within heli

obliged to provide measures to prevent painful patient coolant is directed out of the build experience during MR examination. Such a measure is pipe. Within 1 min the coil curre called a *stimulation monitor*. Works a pro experience during MR examination. Such a measure is *pipe*. Within 1 min the coil current called a *stimulation monitor*. value and the magnetic field is zer porization of the helium is down The temperature of the evapora called a *stimulation monitor*. value and the magnetic field is zero.

Noise Exposure During MR Examination

A conductor carrying a current in the presence of a mag-

to -269° C, and the surrounding en

neutic field w Noise Exposure During MR Examination

The temperature of the evaporate conductor carrying a current in the presence of a mag-

noise the Developmentic field will experience a mechanical force, known cooled accordingly. Th Noise Exposure During MR Examination The temperature of the evap-
A conductor carrying a current in the presence of a mag-
to -269° C, and the surround
netic field will experience a mechanical force, known cooled acco A conductor carrying a current in the presence of a mag-
netic field will experience a mechanical force, known cooled accordingly. There remains the
as the Lorentz force after the Dutch physicist Hen-
when touching covers netic field will experience a mechanical force, known cooled accordingly. There rem
as the Lorentz force after the Dutch physicist Hen-
when touching covers that were drik Antoon Lorentz who described it. The current of r as the Lorentz force after the Dutch physicist Hen-

drik Antoon Lorentz who described it. The current of rating helium. In case of a blockage

up to 600 A flowing through a gradient coil located the helium gas will be fo drik Antoon Lorentz who described it. The current of rating helium. In case of a blockage of
up to 600 A flowing through a gradient coil located the helium gas will be forced into the
within a 1.5 T system will experience up to 600 A flowing through a gradient coil located the helium gas will be forced into the within a 1.5 T system will experience a force on a sin-
there will be the danger of displace
gle wire on the order of 2.827 kN equ within a 1.5 T system will experience a force on a sin-
there will be the danger of displa
gle wire on the order of 2.827 kN equivalent to a weight
that case, the scanner room is to bo
f 635 lb. That force will remain act gle wire on the order of 2.827 kN equivalent to a weight
of 635 lb. That force will remain active for the dura-
are to be planned and exercised to p
tion of the magnetic field gradient being switched on
emergency. There a of 635 lb. That force will remain active for the dura-
tion of the magnetic field gradient being switched on
emergency. There are no know
and will vanish at the time the magnetic field gradi-
have been harmed in case of a tion of the magnetic field gradient being switched on emergency. There are no known cand will vanish at the time the magnetic field gradi-
ent is switched off. The duration of a slice selection
gradient is about 2.5 ms. T and will vanish at the time the magnetic field gradi-

ent is switched off. The duration of a slice selection

gradient is about 2.5 ms. The change from force to
 23.4.6 Gadolinium-Cont

no force is 200 Hz, which genera ent is switched off. The duration of a slice selection
gradient is about 2.5 ms. The change from force to **23.4.6 Gadolinium-Conta**
no force is 200 Hz, which generates a tone between **Contrast Agents an**
a musical a (220 gradient is about 2.5 ms. The change from force to **23.4.6 Gadolinium-Containii**
no force is 200 Hz, which generates a tone between
a musical a (220 Hz) and a g (196 Hz). For a fre-
quency bandwidth of 195 Hz/voxel, the d no force is 200 Hz, which generates a tone between

a musical a (220 Hz) and a g (196 Hz). For a fre-

quency bandwidth of 195 Hz/voxel, the duration of In contrast to iodine-containing

the frequency-encoding gradient wi a musical *a* (220 Hz) and a *g* (196 Hz). For a frequency bandwidth of 195 Hz/voxel, the duration of In contraction the frequency-encoding gradient will be 5.128 ms. The gadolinium switching rate of the gradient would be the frequency-encoding gradient will be 5.128 ms. The gadolinium-covidently sixtiching rate of the gradient would be 97.5 Hz, which considered r
is close to a *G* (98 Hz). Unfortunately the switching of adverse reace
magne g rate of the gradient would be 97.5 Hz, which

o a G (98 Hz). Unfortunately the switching of

field gradients is rarely sinusoidal, and differ-

siasm ev

require different switching frequencies, so the

containing

e

(Quenching)

23.4.5 Safety Issues in Conjunction (gadolinium-diethylenetriamine
 23.4.5 Safety Issues in Conjunction methylamide (Gd-DTPA-BMA

approximately 2 years, of who
 with Loss of Superconductivity oped nephrogenic fibros **23.4.5 Safety Issues in Conjunction** methylamide (Gd-DTPA-BMA)
 approximately 2 years, of whom
 o with Loss of Superconductivity oped nephrogenic fibrosing der
 (Quenching)

The loss of superconductivity of the c **23.4.5 Safety Issues in Conjunction** approximately 2 years, of whom f
with Loss of Superconductivity of the current carrying derno called nephrogenic fibrosis invol
coil of the main magnet is called a *quench*. Currently **With Loss of Superconductivity** oped nephrogenic fibrosing d
 (Quenching) called nephrogenic systemic fibrosis

The loss of superconductivity of the current-carrying joints, eyes, and internal organ

coll of the main m **(Quenching)** called nephrogenic systemic fibro

Nephrogenic systemic fibrosis in

coil of the main magnet is called a *quench*. Currently published in 2000 and has a mort

commercially wailable and relatively easy to han The loss of superconductivity of the current-carrying ints, eyes, and internal organs.

coil of the main magnet is called a *quench*. Currently published in 2000 and has a mort

commercially available and relatively easy

of the switched magnetic field gradients, the ampli-
tudes and switching times of today's gradient systems
are capable of inducing voltages and currents within
helium. After five seconds the pre-
tending work mimic bioche of the switched magnetic field gradients, the ampli-
in the now decaying magnetic fit
ddes and switching times of today's gradient systems to the liquid-helium bath, result
are capable of inducing voltages and currents wi tudes and switching times of today's gradient systems
are capable of inducing voltages and currents within
the lum. After five seconds the pre-
the human body that mimic biochemical voltages usu-
cryostat will be high eno are capable of inducing voltages and currents within helium. After five seconds the press
the human body that mimic biochemical voltages usu-
cryostat will be high enough to act
ally applied to control muscle contraction. the human body that mimic biochemical voltages usu-

ally applied to control muscle contraction. Vendors are disc, as planned in such circums

obliged to provide measures to prevent painful patient coolant is directed out ally applied to control muscle contraction. Vendors are disc, as planned in such circumstance obliged to provide measure so prevent painful patient coolant is directed out of the buildinexperience during MR examination. S Magnetic Resonance Imaging 23.4 MRI – Safety-Relevant Aspects 455
in the now decaying magnetic field will be transferred
to the liquid-helium bath, resulting in vaporization of
cryostat will be high epough to activate the Magnetic Resonance Imaging 23.4 MRI – Safety-Relevant Aspects 455

in the now decaying magnetic field will be transferred

to the liquid-helium bath, resulting in vaporization of

helium. After five seconds the pressure b Magnetic Resonance Imaging 23.4 MRI – Safety-Relevant Aspects
in the now decaying magnetic field will be transferred
to the liquid-helium bath, resulting in vaporization of
helium. After five seconds the pressure build-up Magnetic Resonance Imaging 23.4 MRI – Safety-Relevant Aspects
in the now decaying magnetic field will be transferred
to the liquid-helium bath, resulting in vaporization of
helium. After five seconds the pressure build-up Magnetic Resonance Imaging 23.4 MRI – Safety-Relevant Aspects
in the now decaying magnetic field will be transferred
to the liquid-helium bath, resulting in vaporization of
helium. After five seconds the pressure build-up Magnetic Resonance Imaging 23.4 MRI – Safety-Relevant Aspects 455

in the now decaying magnetic field will be transferred

to the liquid-helium bath, resulting in vaporization of

helium. After five seconds the pressure b Magnetic Resonance Imaging 23.4 MRI – Safety-Relevant Aspects
in the now decaying magnetic field will be transferred
to the liquid-helium bath, resulting in vaporization of
helium. After five seconds the pressure build-up Magnetic Resonance Imaging 23.4 MRI – Safety-Relevant Aspects
in the now decaying magnetic field will be transferred
to the liquid-helium bath, resulting in vaporization of
helium. After five seconds the pressure build-up Magnetic Resonance Imaging 23.4 MRI – Safety-Relevant Aspects
in the now decaying magnetic field will be transferred
to the liquid-helium bath, resulting in vaporization of
helium. After five seconds the pressure build-up in the now decaying magnetic field will be transferred
to the liquid-helium bath, resulting in vaporization of
helium. After five seconds the pressure build-up in the
cryostat will be high enough to activate the blowout
d in the now decaying magnetic field will be transferred
to the liquid-helium bath, resulting in vaporization of
helium. After five seconds the pressure build-up in the
cryostat will be high enough to activate the blowout
d in the now decaying magnetic field will be transferred
to the liquid-helium bath, resulting in vaporization of
helium. After five seconds the pressure build-up in the
cryostat will be high enough to activate the blowout
d to the liquid-helium bath, resulting in vaporization of
helium. After five seconds the pressure build-up in the
cryostat will be high enough to activate the blowout
disc, as planned in such circumstances. The evaporating
 helium. After five seconds the pressure build-up in the
cryostat will be high enough to activate the blowout
disc, as planned in such circumstances. The evaporating
coolant is directed out of the building through a *quenc* cryostat will be high enough to activate the blowout
disc, as planned in such circumstances. The evaporating
coolant is directed out of the building through a *quench*
pipe. Within 1 min the coil currents have reached a z disc, as planned in such circumstances. The evaporating
coolant is directed out of the building through a *quench*
pipe. Within 1 min the coil currents have reached a zero
value and the magnetic field is zero. After 2 min coolant is directed out of the building through a *quench*
pipe. Within 1 min the coil currents have reached a zero
value and the magnetic field is zero. After 2 min, the va-
porization of the helium is down to a negligib pipe. Within 1 min the coil currents have reached a zero
value and the magnetic field is zero. After 2 min, the va-
porization of the helium is down to a negligible level.
The temperature of the evaporating helium is clos value and the magnetic field is zero. After 2 min, the va-
porization of the helium is down to a negligible level.
The temperature of the evaporating helium is close
to -269° C, and the surrounding environment will be porization of the helium is down to a negligible level.
The temperature of the evaporating helium is close
to -269°C , and the surrounding environment will be
cooled accordingly. There remains the danger of burns
w to -269° C, and the surrounding environment will be cooled accordingly. There remains the danger of burns when touching covers that were affected by the evaporating helium. In case of a blockage of the quench pipes, t cordingly. There remains the danger of burns
ching covers that were affected by the evapo-
ium. In case of a blockage of the quench pipes,
n gas will be forced into the scanner room and
l be the danger of displacement of o rating helium. In case of a blockage of the quench pipes,
the helium gas will be forced into the scanner room and
there will be the danger of displacement of oxygen. In
that case, the scanner room is to be evacuated. Measu the helium gas will be forced into the scanner room and
there will be the danger of displacement of oxygen. In
that case, the scanner room is to be evacuated. Measures
are to be planned and exercised to prepare for such an

ent tasks require different switching frequencies, so the

final noise is rarely considered harmonic. It is required

imaging [23.51, 52].

that the noise level for a patient remain below 99 dB(A) entists at AKH in

with final noise is rarely considered harmonic. It is required imaging [23.51, 52]. In 2
that the noise level for a patient remain below 99 dB(A) entists at AKH in Vienna
with or without ear-protective devices. No vendor is al that the noise level for a patient remain below 99 dB(A) entists at AKH in Vienna publ
with or without ear-protective devices. No vendor is al-
lowed to introduce a scanner to the market capable of went MR angiography usi with or without ear-protective devices. No vendor is al-

lowed to introduce a scanner to the market capable of went MR angiography using a given

generating more than 140 dB(A).

23.4.5 Safety Issues in Conjunction

with Nowed to introduce a scanner to the market capable of went MR angiography using
generating more than 140 dB(A). (gadolinium-diethylenetriamine
methylamide (Gd-DTPA-BMA
approximately 2 years, of who
with Loss of Supercond there will be the danger of displacement of oxygen. In
that case, the scanner room is to be evacuated. Measures
are to be planned and exercised to prepare for such an
emergency. There are no known cases where humans
have b that case, the scanner room is to be evacuated. Measures
are to be planned and exercised to prepare for such an
emergency. There are no known cases where humans
have been harmed in case of a quench.
23.4.6 Gadolinium-Cont are to be planned and exercised to prepare for such an
emergency. There are no known cases where humans
have been harmed in case of a quench.
23.4.6 Gadolinium-Containing
Contrast Agents and NSF
In contrast to iodine-c emergency. There are no known cases where humans
have been harmed in case of a quench.
 23.4.6 Gadolinium-Containing
 Contrast Agents and NSF

In contrast to iodine-containing contrast media,

gadolinium-containing MR have been harmed in case of a quench.
 23.4.6 Gadolinium-Containing
 Contrast Agents and NSF

In contrast to iodine-containing contrast media,

gadolinium-containing MR contrast agents were long

considered non-nephrot **23.4.6 Gadolinium-Containing**
 Contrast Agents and NSF

In contrast to iodine-containing contrast media,

gadolinium-containing MR contrast agents were long

considered non-nephrotoxic with very low risk of

adverse rea **23.4.6 Gadolinium-Containing**
 Contrast Agents and NSF

In contrast to iodine-containing contrast media,

gadolinium-containing MR contrast agents were long

considered non-nephrotoxic with very low risk of

adverse re **Contrast Agents and NSF**
In contrast to iodine-containing contrast media,
gadolinium-containing MR contrast agents were long
considered non-nephrotoxic with very low risk of
adverse reactions or other complications. The In contrast to iodine-containing contrast media,
gadolinium-containing MR contrast agents were long
considered non-nephrotoxic with very low risk of
adverse reactions or other complications. The enthu-
siasm even triggered In contrast to iodine-containing contrast media,
gadolinium-containing MR contrast agents were long
considered non-nephrotoxic with very low risk of
adverse reactions or other complications. The enthu-
siasm even triggere gadolinium-containing MR contrast agents were long
considered non-nephrotoxic with very low risk of
adverse reactions or other complications. The enthu-
siasm even triggered suggestions to use gadolinium-
containing contra considered non-nephrotoxic with very low risk of
adverse reactions or other complications. The enthu-
siasm even triggered suggestions to use gadolinium-
containing contrast agents in conjunction with x-ray
imaging [23.51, adverse reactions or other complications. The enthu-
siasm even triggered suggestions to use gadolinium-
containing contrast agents in conjunction with x-ray
imaging [23.51, 52]. In 2006, a group of sci-
entists at AKH in siasm even triggered suggestions to use gadolinium-
containing contrast agents in conjunction with x-ray
imaging [23.51, 52]. In 2006, a group of sci-
entists at AKH in Vienna published a study of
nine end-stage renal dis containing contrast agents in conjunction with x-ray
imaging [23.51, 52]. In 2006, a group of sci-
entists at AKH in Vienna published a study of
nine end-stage renal disease patients who under-
went MR angiography using a imaging [23.51, 52]. In 2006, a group of scientists at AKH in Vienna published a study of nine end-stage renal disease patients who underwent MR angiography using a gadolinium chelate (gadolinium-diethylenetriaminepentaac entists at AKH in Vienna published a study of
nine end-stage renal disease patients who under-
went MR angiography using a gadolinium chelate
(gadolinium-diethylenetriaminepentaacetic acid bis-
methylamide (Gd-DTPA-BMA)) o nine end-stage renal disease patients who underwent MR angiography using a gadolinium chelate (gadolinium-diethylenetriaminepentaacetic acid bis-
methylamide (Gd-DTPA-BMA)) over a period of approximately 2 years, of whom f went MR angiography using a gadolinium chelate (gadolinium-diethylenetriaminepentaacetic acid bis-
methylamide (Gd-DTPA-BMA)) over a period of approximately 2 years, of whom five patients devel-
oped nephrogenic fibrosing (gadolinium-diethylenetriaminepentaacetic acid bis-
methylamide (Gd-DTPA-BMA)) over a period of
approximately 2 years, of whom five patients devel-
oped nephrogenic fibrosing dermopathy (NFD), also
called nephrogenic syste methylamide (Gd-DTPA-BMA)) over a period of
approximately 2 years, of whom five patients devel-
oped nephrogenic fibrosing dermopathy (NFD), also
called nephrogenic systemic fibrosis involves fibrosis of skin,
joints, eyes approximately 2 years, of whom five patients devel-
oped nephrogenic fibrosing dermopathy (NFD), also
called nephrogenic systemic fibrosis involves fibrosis of skin,
Nephrogenic systemic fibrosis involves fibrosis of skin, oped nephrogenic fibrosing dermopathy (NFD), also
called nephrogenic systemic fibrosis (NSF) [23.53, 54].
Nephrogenic systemic fibrosis involves fibrosis of skin,
joints, eyes, and internal organs. The disease was first
pu called nephrogenic systemic fibrosis (NSF) [23.53, 54].
Nephrogenic systemic fibrosis involves fibrosis of skin,
joints, eyes, and internal organs. The disease was first
published in 2000 and has a mortality rate of 5%. Si Aggretic Resonance imaging [23.4 MR - 5drty-Relevant Aspects 465

in the now decouply magnetic field will be transferred in the norm of the model in the model of the proposition of the behach of the proposition of the pro

Medical Imaging

23.5 MRI – Pictures of the Future

For the past 20 years it has frequently been predicted

that developments within the area of magnetic reso-Medical Imaging
 23.5 MRI – Pictures of the Future

For the past 20 years it has frequently been predicted

that developments within the area of magnetic reso-

nance and magnetic resonance imaging would reach the signa Medical Imaging
 23.5 MRI – Pictures of the Future

For the past 20 years it has frequently been predicted

that developments within the area of magnetic reso-

nance and magnetic resonance imaging would reach

The sign Medical Imaging
 23.5 MRI – Pictures of the Future

For the past 20 years it has frequently been predicted

The signal gain when working that developments within the area of magnetic resonance

maturity, and further pro **23.5 MRI – Pictures of the Future**

For the past 20 years it has frequently been predicted 23.5.2 RF Technology

that developments within the area of magnetic reso-

nance and magnetic resonance imaging would reach The s **Example 12**
 Example 13.5 MRI – Pictures of the Future

For the past 20 years it has frequently been predicted

123.5.2 RF Technology

that developments within the area of magnetic reso-

nance and magnetic resonance i **Puter Sof the Future**
 Putures of the Future

For the past 20 years it has frequently been predicted

The signal gain when working that developments within the area of magnetic reso-

nance and magnetic resonance imagi times. **23.5 MRI – Pictures of the Future**

For the past 20 years it has frequently been predicted

that developments within the area of magnetic reso-

nance and magnetic resonance imaging would reach

maturity, and further pro For the past 20 years it has frequently been predicted

that developments within the area of magnetic reso-

nance and magnetic resonance imaging would reach

maturity, and further progress would be slow and only

field s

incremental – similar to the development in x-ray com-

puted tomography. So far we are still waiting for those

tion with the utilization of spatial

times.
 23.5.1 Magnetic Field Strength
 23.5.1 Magnetic Field Stren puted tomography. So far we are still waiting for those

innextigation of spatial

imes.

23.5.1 Magnetic Field Strength

23.5.1 Magnetic Field Strength

About two-thirds of the market is currently work-

but the market i **23.5.1 Magnetic Field Strength**
 alternation spatial information would involve
 23.5.1 Magnetic Field Strength

About two-thirds of the market is currently work-

imatrices for *parallel imaging* is a

high with magn **23.5.1 Magnetic Field Strength**

spatial information would

institute that the market is currently work-

ing with magnetic field strength of 1.5 T. High-field systems are currently prep

systems with magnetic field stre **23.5.1 Magnetic Field Strength**

ditional phase-encoding steps. T

matrices for *parallel imaging* is a

ing with magnetic field strength of 1.5 T. High-field systems are currently prepared to

systems with remaining de-About two-thirds of the market is currently work-
ing with magnetic field strength of 1.5 T. High-field systems are currently prepared to
systems with magnetic field strength of 3.0 T have 128 independent receiver channel humans.

For the past 20 years it has frequently been predicted **23.5.2 RF Technology**

that developments within the area of magnetic reso-

maturity, and further progress would be slow and only field strength (Fig. 23.12) is supp that developments within the area of magnetic reso-
nance and magnetic resonance imaging would reach
maturity, and further progress would be slow and only
field strength (Fig. 23.12) is supple
incremental – similar to the mance and magnetic resonance imaging would reach
maturity, and further progress would be slow and only field strength (Fig. 23.12) is suppler
incremental – similar to the development in x-ray com-
measures to reduce measu maturity, and further progress would be slow and only
incremental – similar to the development in x-ray com-
measures to reduce measure
puted tomography. So far we are still waiting for those tion with the utilization of 23.5.2 RF Technology
The signal gain when working with higher magnetic
field strength (Fig. 23.12) is supplemental and support **23.5.2 RF Technology**
The signal gain when working with higher magnetic
field strength (Fig. 23.12) is supplemental and supports
measures to reduce measurement times in conjunc-
tion with the utilization of spatial infor **23.5.2 RF Technology**
The signal gain when working with higher magnetic
field strength (Fig. 23.12) is supplemental and supports
measures to reduce measurement times in conjunc-
tion with the utilization of spatial inform **23.5.2 RF Technology**
The signal gain when working with higher magnetic
field strength (Fig. 23.12) is supplemental and supports
measures to reduce measurement times in conjunc-
idistributed coil matrices. Otherwise, mea **23.5.2 RF Technology**
The signal gain when working with higher magnetic
field strength (Fig. 23.12) is supplemental and supports
measures to reduce measurement times in conjunc-
tion with the utilization of spatial infor **23.5.2 RF Technology**
The signal gain when working with higher magnetic
field strength (Fig. 23.12) is supplemental and supports
measures to reduce measurement times in conjunc-
tion with the utilization of spatial infor **23.5.2 RF Technology**
The signal gain when working with higher magnetic
field strength (Fig. 23.12) is supplemental and supports
measures to reduce measurement times in conjunc-
tion with the utilization of spatial infor **23.5.2 RF Technology**
The signal gain when working with higher magnetic
field strength (Fig. 23.12) is supplemental and supports
measures to reduce measurement times in conjunc-
tion with the utilization of spatial infor **23.5.2 RF Technology**
The signal gain when working with higher magnetic
field strength (Fig. 23.12) is supplemental and supports
measures to reduce measurement times in conjunc-
tion with the utilization of spatial infor **23.5.2 RF Technology**
The signal gain when working with higher magnetic
field strength (Fig. 23.12) is supplemental and supports
measures to reduce measurement times in conjunc-
tion with the utilization of spatial infor **23.5.2 RF Technology**
The signal gain when working with higher magnetic
field strength (Fig. 23.12) is supplemental and supports
measures to reduce measurement times in conjunc-
tion with the utilization of spatial infor The signal gain when working with higher magnetic
field strength (Fig. 23.12) is supplemental and supports
measures to reduce measurement times in conjunc-
tion with the utilization of spatial information from
distributed The signal gain when working with higher magnetic
field strength (Fig. 23.12) is supplemental and supports
measures to reduce measurement times in conjunc-
tion with the utilization of spatial information from
distributed field strength (Fig. 23.12) is supplemental and supports
measures to reduce measurement times in conjunc-
tion with the utilization of spatial information from
distributed coil matrices. Otherwise, measurement of
spatial i measures to reduce measurement times in conjunc-
tion with the utilization of spatial information from
distributed coil matrices. Otherwise, measurement of
spatial information would involve time-consuming ad-
ditional phas tion with the utilization of spatial information from
distributed coil matrices. Otherwise, measurement of
spatial information would involve time-consuming ad-
ditional phase-encoding steps. The ability to use coil
matrice spatial information would involve time-consuming ad-
ditional phase-encoding steps. The ability to use coil
matrices for *parallel imaging* is a function of the num-
ber of coils and coil profiles. Modern commercial MR
sys matrices for *parallel imaging* is a function of the num-
ber of coils and coil profiles. Modern commercial MR
systems are currently prepared to serve coils of up to
128 independent receiver channels, and the first proto-
 Part C Medical imaging
 Figure 10.1 Control in the Culture Soft the Future Register estate that the control in the case of magnetic resonance with the case of magnetic resonance imaging would reach The signal gain

Fig. 23.12 Image of a T_2 -weighted
axial cut of the brain of a patient
with multiple sclerosis, performed
on a 7.0 T system (courtesy of the Fig. 23.12 Image of a T_2 -weighted
axial cut of the brain of a patient
with multiple sclerosis, performed
on a 7.0 T system (courtesy of the
University Hospitals of New York) Fig. 23.12 Image of a T₂-weighted
axial cut of the brain of a patient
with multiple sclerosis, performed
on a 7.0 T system (courtesy of the
University Hospitals of New York) Fig. 23.12 Image of a T₂-weighted
axial cut of the brain of a patient
with multiple sclerosis, performed
on a 7.0 T system (courtesy of the
University Hospitals of New York) Fig. 23.12 Image of a *T*₂-weighted
axial cut of the brain of a patient
with multiple sclerosis, performed
on a 7.0 T system (courtesy of the
University Hospitals of New York)

- Motion correction, registration, and mapping of
anatomic structures
Mapping and visualization of different (additional)
information (parametric imaging)
Perfusion measurements without contrast agents Motion correction, registration, and mapping of PET system
anatomic structures
matomic structures within the
Mapping and visualization of different (additional) MR-PET sy
information (parametric imaging) a question of
Perf Motion correction, registration, and mapping of PET system

anatomic structures

Mapping and visualization of different (additional) MR-PET

information (parametric imaging) a question

Perfusion measurements without contr
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- Motion correction, registration, and mapping of PET system as

anatomic structures

 Mapping and visualization of different (additional) MR-PET system

information (parametric imaging) and visualization of different (ad Magnetic Resonance Imaging

Motion correction, registration, and mapping of PET system as a hybrid tech

anatomic structures

Mapping and visualization of different (additional) MR-PET system as a routine

information (par Motion correction, registration, and mapping of PET

anatomic structures

Mapping and visualization of different (additional) MR-

information (parametric imaging) a que-

Perfusion measurements without contrast agents mod • Motion correction, registration, and mapping of PET system as a hybrid techn

anatomic structures

• Mapping and visualization of different (additional) MR-PET system as a routine

information (parametric imaging)

• Per anatomic structures

Mapping and visualization of different (additional) MR-PET system as a routine

information (parametric imaging)

a question of time. MR-PET c

Perfusion measurements without contrast agents modalities Mapping and visualization of different (additional) MR-PET system a
information (parametric imaging) a question of time.
Perfusion measurements without contrast agents modalities. The sure
(arterial spin labeling (ASL)) ti The exception of the scanner divide provide in the scanner will server that the current in the current state of the current state of the current state of the current standardiza-

In general it can be stated that the curre
-

• MR angiography without contrast agents (native) • PET to provide biochemical • Time-resolved omtrast-enhanced MR angiography • Potential scientific applic:

(time-resolved imaging with interleaved stochastic trajectories • Time-resolved contrast-enhanced MR angiography

(time-resolved imaging with interleaved stochastic

trajectories (TWIST) and time-resolved imaging of

contrast kinetics (TRICKS))

• MR data acquisition in the presence of (time-resolved imaging with interleaved stochastic

trajectories (TWIST) and time-resolved imaging of

OMR data acquisition in the presence of a continu-

MR data acquisition in the presence of a continu-

ME dimensional trajectories (TWIST) and time-resolved imaging of

contrast kinetics (TRICKS) (real-time information corre

MR data acquisition in the presence of a continu-

oriental time information with di

ously moving patient table (CONTERT CONTECT (TRICKS)

• MR data acquisition in the presence of a continuous

original could matrix with continuous table movement, MDS – (3-D-CSI), and high-resolution

move during scan).

• CONTI), three dimensional c • MR data acquisition in the presence of a continu-

ously moving patient table (TimCT – total imaging (DTI), three dim

matrix with continuous table movement, MDS – (3-D-CSI), and hi

move during scan).

In general it can ously moving patient table (TimCT – total imaging (DTI), three dimensional chem

matrix with continuous table movement, MDS – (3-D-CSI), and high-resolution of

In general it can be stated that the current standardiza-
 • matrix with continuous table movement, MDS – (3-D-CSI), and high-resolution s

In general it can be stated that the current standardiza-
 In general it can be stated that the current standardiza-
 IDENTIFY CONTEX CONTEX move during scan).

In general it can be stated that the current standardiza-

in anatomical structures within anatomical structures within anatomical structures within anatomical structures and in shorter times during whi In general it can be stated that the current standardiza-

tion and automatization is streamlining the workflow,

resulting in shorter times during which patient has to

remain in the scanner, while simultaneously ensuing
 In general it can be stated that the current standardiza-

ion and automatization is streamlining the workflow,

resulting in shorter times during which patient has to

remain in the scanner, while simultaneously ensuring
 tion and automatization is streamlining the workflow,

resulting in shorter times during which patient has to

remain in the scanner, while simultaneously ensuring

reproducible results; e.g., for brain studies, slice orie resulting in shorter times during which patient has to

remain in the scanner, while simultaneously ensuring

reproducible results; e.g., for brain studies, slice orienta-

tions and coverage are recommended in specific gu remain in the scanner, while simultaneously ensuring

reproducible results; e.g., for brain studies, slice orienta-

tions and coverage are recommended in specific guide-

lines. In modern current systems a three-dimension reproducible results; e.g., for brain studies, slice orienta-
tions and coverage are recommended in specific guide-
lines. In modern current systems a three-dimensional
diation necrosis
localizer is usually initiated autom tions and coverage are recommended in specific guide-

lines. In modern current systems a three-dimensional

localizer is usually initiated automatically after the packing and diation necrosis

localizer is usually initiat lines. In modern current systems a three-dimensional

localizer is usually initiated automatically after the pa-

tient has been positioned. Current modern systems are

immediately able to adjust the next imaging protocol
 localizer is usually initiated automatically after the packing the test intent has been positioned. Current modern systems are

immediately able to adjust the next imaging protocol

Staging, restaging, and there

with resp tient has been positioned. Current modern systems are

immediately able to adjust the next imaging protocol

with respect to angulation and requested coverage in

artic oncology

automatically analyzing the 3-D-localizer f immediately able to adjust the next imaging protocol

with respect to angulation and requested coverage in

atric oncology

automatically analyzing the 3-D-localizer for anatom-

• Whole-body tumor staging an

cical struct with respect to angulation and requested coverage in atric oncology
automatically analyzing the 3-D-localizer for anatom-
ical structures and landmarks. Similar approaches are • Myocardial in
currently already available to matically analyzing the 3-D-localizer for anatom-

structures and landmarks. Similar approaches are

the Myocardial infarction: differently already available to automatically position re-

then the structure of selected el ical structures and landmarks. Similar approaches are

currently already available to automatically position re-

equired silices and volumes for studying the knee of

a patient. For imaging the heart, current modern syste currently already available to automatically position re-

elemention and stunning.

quired slices and volumes for studying the knee of

a patient. For imaging the heart, current modern systems There are a few challenges t quired slices and volumes for studying the knee of
a patient. For imaging the heart, current modern systems
utilize the localizer to select the recommended short-
grating a PET system within a ME
axis or long-axis views au

a patient. For imaging the heart, current modern systems There are a few challenges to
utilize the localizer to select the recommended short-
graing a PET system within a
axis or long-axis views automatically. In imaging t utilize the localizer to select the recommended short-

axis or long-axis views automatically. In imaging the

spine, a single mouse click can indicate to the system

which intervertebral disc space is to be evaluated, and spine, a single mouse click can indicate to the system
which intervertebral disc space is to be evaluated, and
the system will automatically adjust the imaging plane
to have the same angulation as the disc.
The number of s the system will automatically adjust the imaging plane
to have the same angulation as the disc.
The number of selected elements of a surface coil
will affect the coverage as well as the image noise. If
more coil elements a The number of selected elements of a surface coil

will affect the coverage as well as the image noise. If

more coil elements selected than needed, the im-

age noise is unnecessarily increased. If the number of

coil ele will affect the coverage as well as the image noise. If
more coil elements are selected than needed, the im-
age noise is unnecessarily increased. If the number of
coil elements selected is too low, the coverage neces-
sar

• Motion correction, registration, and mapping of PET system as a hybrid technology

• Motion correction, registration, and mapping of PET system as a hybrid technology

• Mapping and visualization of different (additional • Motion correction, registration, and mapping of PET system as a hybrid te anatomic structures

• Mapping and visualization of different (additional) MR-PET system as a routi

information (parametric imaging) and visualiz • Motion correction, registration, and mapping of PET system as a hybrid technol anatomic structures

• Mapping and visualization of different (additional) MR-PET system as a routine climformation (parametric imaging) and • Motion correction, registration, and mapping of PET system as a hybrid tech anatomic structures

• Mapping and visualization of different (additional) MR-PET system as a routing information (parametric imaging) a questio Motion correction, registration, and mapping of PET system as a hybrid technology

anatomic structures

Mapping and visualization of different (additional) MR-PET system as a routine clinical

information (parametric imagi Magnetic Resonance Imaging 23.5 MRI – Pictures of the Future 457

PET system as a hybrid technology has also surfaced

within the MR community. The introduction of an

MR-PET system as a routine clinical modality is only
 Magnetic Resonance Imaging 23.5 MRI – Pictures of the Future

PET system as a hybrid technology has also surfaced

within the MR community. The introduction of an

MR-PET system as a routine clinical modality is only

a q Magnetic Resonance Imaging 23.5 MRI – Pictures of the Future

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 Magnetic Resonance Imaging 23.5 MRI – Pictures of the Future

PET system as a hybrid technology has also surfaced

within the MR community. The introduction of an

MR-PET system as a routine clinical modality is only

a q Example 123.5 MRI – Pictures of the Future

Potential science of the Future

Postem as a hybrid technology has also surfaced

in the MR community. The introduction of an

-PET system as a routine clinical modality is only **FRIT system as a hybrid technology has also surfaced**
within the MR community. The introduction of an
MR-PET system as a routine clinical modality is only
a question of time. MR-PET combines complementary
modalities. The F system as a hybrid technology has also surfaced

in the MR community. The introduction of an

-PET system as a routine clinical modality is only

uestion of time. MR-PET combines complementary

dalities. The superior ab **PET system as a hybrid technology has also surfaced**
within the MR community. The introduction of an
MR-PET system as a routine clinical modality is only
a question of time. MR-PET combines complementary
modalities. The s in the MR community. The introduction of an
-PET system as a routine clinical modality is only
the estion of time. MR-PET combines complementary
dalities. The superior ability of MR to display soft
the contrast is combined -PET system as a routine clinical modality is only

lestion of time. MR-PET combines complementary

lalities. The superior ability of MR to display soft

ue contrast is combined with the unique feature of

¹ to provide a question of time. MR-PET combines complementary

modalities. The superior ability of MR to display soft

tissue contrast is combined with the unique feature of

PET to provide biochemical information.

Potential scienti

-
- decribatives. The superior ability of MR to display soft
the contrast is combined with the unique feature of
The provide biochemical information.
Potential scientific applications are:
Simultaneous activation studies by PE FIRE to provide biochemical information.

FORE TO provide biochemical information.

Potential scientific applications are:

• Simultaneous activation studies by PET and fMRI

(real-time information correlation)

• PET comb F to provide biochemical information.

Potential scientific applications are:

Simultaneous activation studies by PET and fMRI

(real-time information correlation)

PET combination with diffusion tensor imaging

(DTI), thr Simultaneous activation studies by PET and fMRI
(real-time information correlation)
PET combination with diffusion tensor imaging
(DTI), three dimensional chemical shift imaging
(3-D-CSI), and high-resolution structural MR (real-time information correlation)

• PET combination with diffusion tensor imaging

(DTI), three dimensional chemical shift imaging

(3-D-CSI), and high-resolution structural MRI

• Dynamics of distribution of pharmaceut PET combination with diffusion tensor imagir

(DTI), three dimensional chemical shift imagir

(3-D-CSI), and high-resolution structural MRI

Dynamics of distribution of pharmaceutical pro

ucts within anatomical structures (DTI), three dimensional chemical shift imaging

(3-D-CSI), and high-resolution structural MRI

■ Dynamics of distribution of pharmaceutical products within anatomical structures

■ Development and evaluation of cell the
- (3-D-CSI), and high-resolution structural MRI
Dynamics of distribution of pharmaceutical prod-
ucts within anatomical structures
Development and evaluation of cell therapy (stem-
cell migration tracking and differentiation • Dynamics of distribution of pharmaceutical products within anatomical structures

• Development and evaluation of cell therapy (stem-

cell migration tracking and differentiation).

Potential clinical applications are:

- ucts within anatomical structures
Development and evaluation of cell therapy (s
cell migration tracking and differentiation).
Potential clinical applications are:
Differential diagnosis of recurrent tumors versu
diation ne

-
-
-
-
- Development and evaluation of cell therapy (stem-

cell migration tracking and differentiation).

Potential clinical applications are:

 Differential diagnosis of recurrent tumors versus ra-

diation necrosis

 Early d dell migration tracking and differentiation).

Potential clinical applications are:

• Differential diagnosis of recurrent tumors versus radiation necrosis

• Early diagnosis of Alzheimer's disease – improved

prognosis du Potential clinical applications are:

Differential diagnosis of recurrent tumors versus ra-

diation necrosis

Early diagnosis of Alzheimer's disease – improved

prognosis due to potentially earlier medication

Staging, re

• Differential diagnosis of recurrent tumors versus radiation necrosis
• Early diagnosis of Alzheimer's disease – improved
prognosis due to potentially earlier medication
• Staging, restaging, and therapy monitoring in ped • Differential diagnosis of recurrent tumors versus radiation necrosis
• Early diagnosis of Alzheimer's disease – improved
prognosis due to potentially earlier medication
• Staging, restaging, and therapy monitoring in ped diation necrosis

• Early diagnosis of Alzheimer's disease – improved

prognosis due to potentially earlier medication

• Staging, restaging, and therapy monitoring in pedi-

atric oncology

• Whole-body tumor staging and

Medical Imaging
(Fig. 23.13). Beneficial for this development has been As is obvious from the above,
the early introduction of MR systems with a larger bore neers as well as physicians and teer
an extremely exciting and in Medical Imaging

(Fig. 23.13). Beneficial for this development has been As is obvious from the above,

the early introduction of MR systems with a larger bore neers as well as physicians and ted

diameter.

23.5.5 Therapo diameter. Medical Imaging

(Fig. 23.13). Beneficial for this development has been

the early introduction of MR systems with a larger bore

neers as v

diameter.

23.5.5 Theranostics –

Therapy Under Image Guidance

Compare Furth

C maging

3). Beneficial for this development has been

introduction of MR systems with a larger bore

an extremely exciting

an extremely exciting

all these potential ne
 Therapy Under Image Guidance
 Therapy Under Imag

Medical Imaging

(Fig. 23.13). Beneficial for this development has been

the early introduction of MR systems with a larger bore neers as well as physicians and tech

diameter.
 23.5.5 Theranostics -
 23.5.5 Theranostic Example 18
 IFORE 18.23.13). Beneficial for this development has been as well as physicians and the early introduction of MR systems with a larger bore neers as well as physicians and an extremely exciting and interedu (Fig. 23.13). Beneficial for this development has been

the early introduction of MR systems with a larger bore

and extremely exciting and intereducer.
 23.5.5 Theranostics –
 23.5.5 Theranostics –
 23.5.5 Theranost (Fig. 23.13). Beneficial for this development has been

the early introduction of MR systems with a larger bore

in extremely exciting and interesting

all these potential new and novel deve

an extremely exciting and int (Fig. 23.13). Beneficial for this development has been

the early introduction of MR systems with a larger bore

diameter.
 23.5.5 Theranostics -
 23.5.5 Theranostics -
 23.5.5 Theranostics -
 23.5.5 Theranostics diameter.
 23.5.5 Theranostics
 20.5 Theranostics
 20.5 Theranostics
 20.6 T diameter.
 23.5.5 Theranostics –
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