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John P. Greenwood
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Cardiovascular Magnetic Resonance

Physics for Clinicians
Pocket Guide

David Broadbent
Ananth Kidambi
John Biglands

Foreword

Cardiovascular Magnetic Resonance physics is an ever expanding field with new innovations progressing into clinical practice on a regular basis. This pocket guide aims to provide a day-to-day companion for those working in CMR. The aim is to provide a short overview of the physics underlying the main investigations used in CMR in a format that is accessible to both clinicians and scientists. The guide opens with fundamental principles of MRI then progresses through standard imaging techniques and on to techniques specific to CMR.

This PDF book has interactive index buttons that are compatible with most desktop PDF readers, the Apple iBooks app and Adobe Reader for Android.

Acknowledgements

We are grateful to Dr John Ridgway for his invaluable assistance and support with the production of this guide.
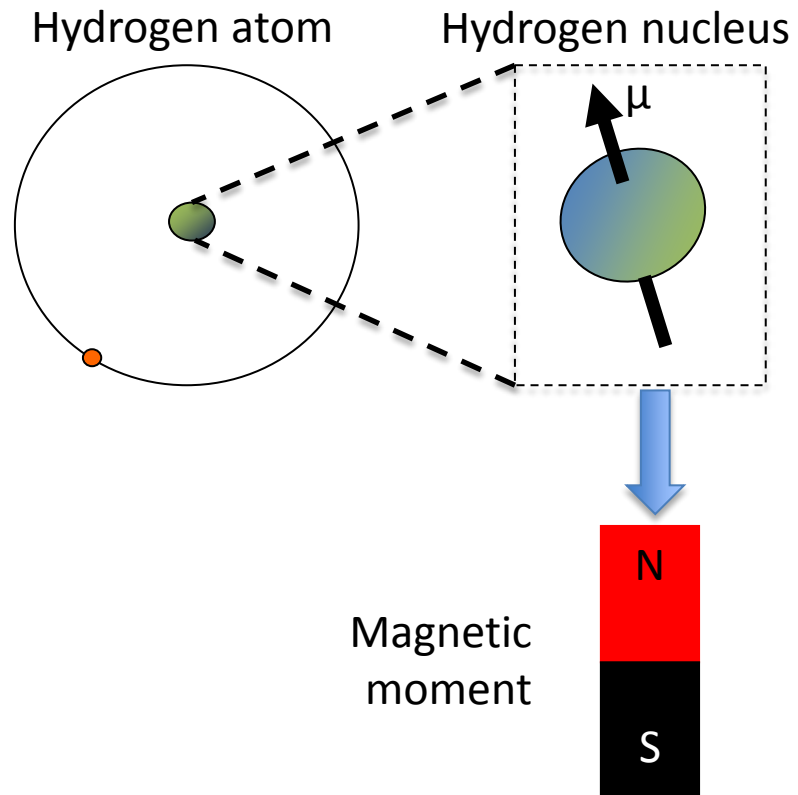
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Contents

The Nuclear Magnetic Moment

- The origin of the MRI signal is the hydrogen nuclei, which has an inherent property called 'spin' giving the nucleus a small magnetic moment μ.

Hydrogen atom   Hydrogen nucleus

μ

Magnetic moment

N

S

The External Magnetic Field, B_0

- In the presence of a strong magnetic field (B_0) in the z-direction spins:
a) exhibit a small preference to align with B_0 giving a net magnetisation M_z aligned with B_0.
b) precess about B_0 at the resonant frequency ω_0 (proportional to the strength of B_0), but with random phase, giving a net transverse magnetisation $M_{xy} = 0$.

a) M_z z B_0

M_z

b) M_{xy} z B_0

$\omega_0 = \gamma B_0$

$M_{xy} = 0$ y

x
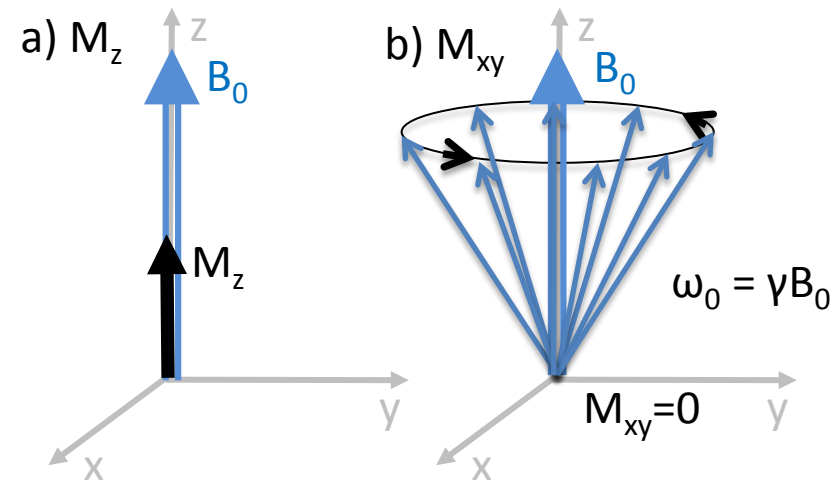
(γ is the gyromagnetic ratio.)

The Transmitted RF Pulse

- A magnetic field applied perpendicular to B_0 and oscillating at the resonant frequency ω_0, is applied. This is the radiofrequency field, B_1. The net magnetisation precesses about this field such that a component of the net magnetisation tips into the transverse plane.

- For clinical MR systems the resonant frequency falls in the radiofrequency region of the electromagnetic spectrum.
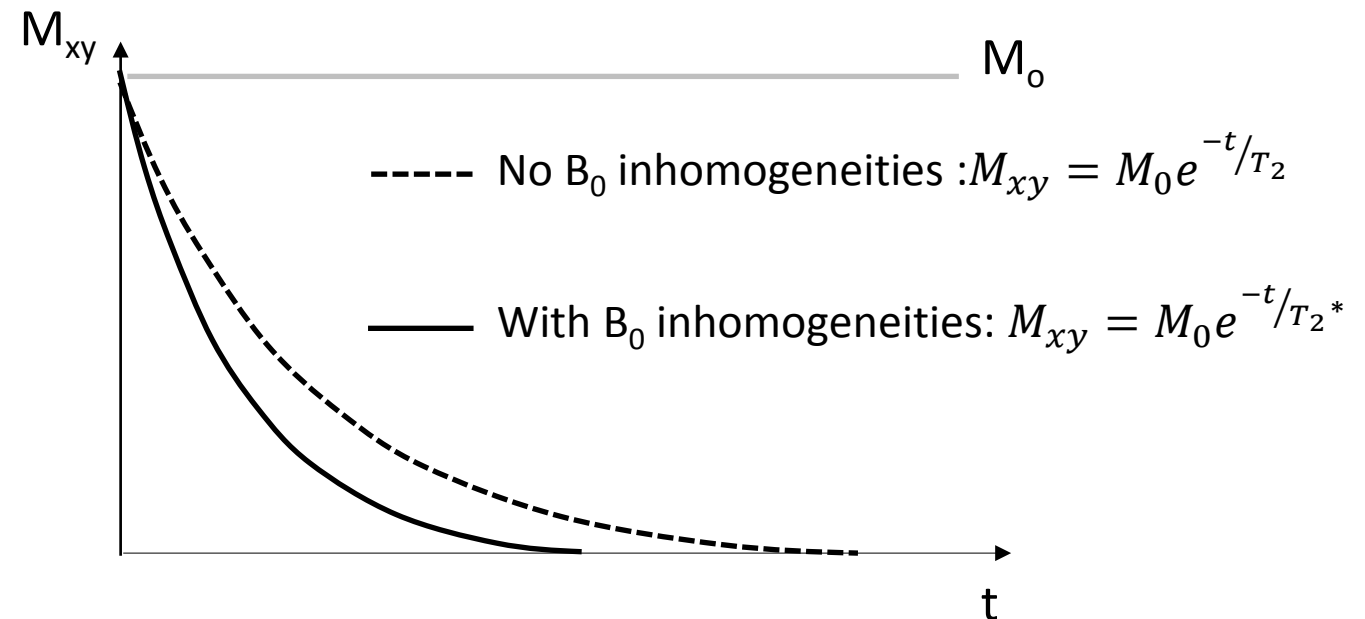
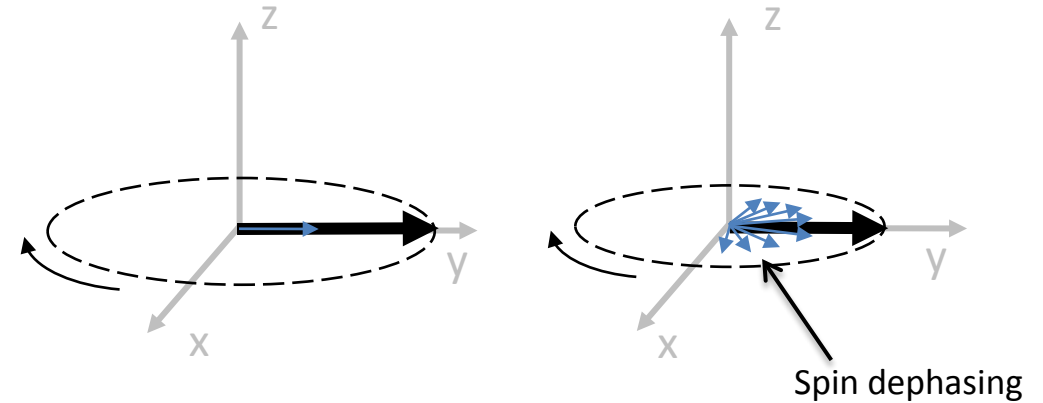## The Received RF Signal

- After B_1 is turned off the transverse component of the net magnetisation, M_{xy}, continues to precess about B_0. This moving magnetic moment will induce a voltage in a conducting coil placed nearby (the RF receive coil). This voltage is the measured MR signal.

B_0

M_z

M_{xy}

B_1

Receive RF coil

M_{xy}

Measured signal (free induction decay, FID)
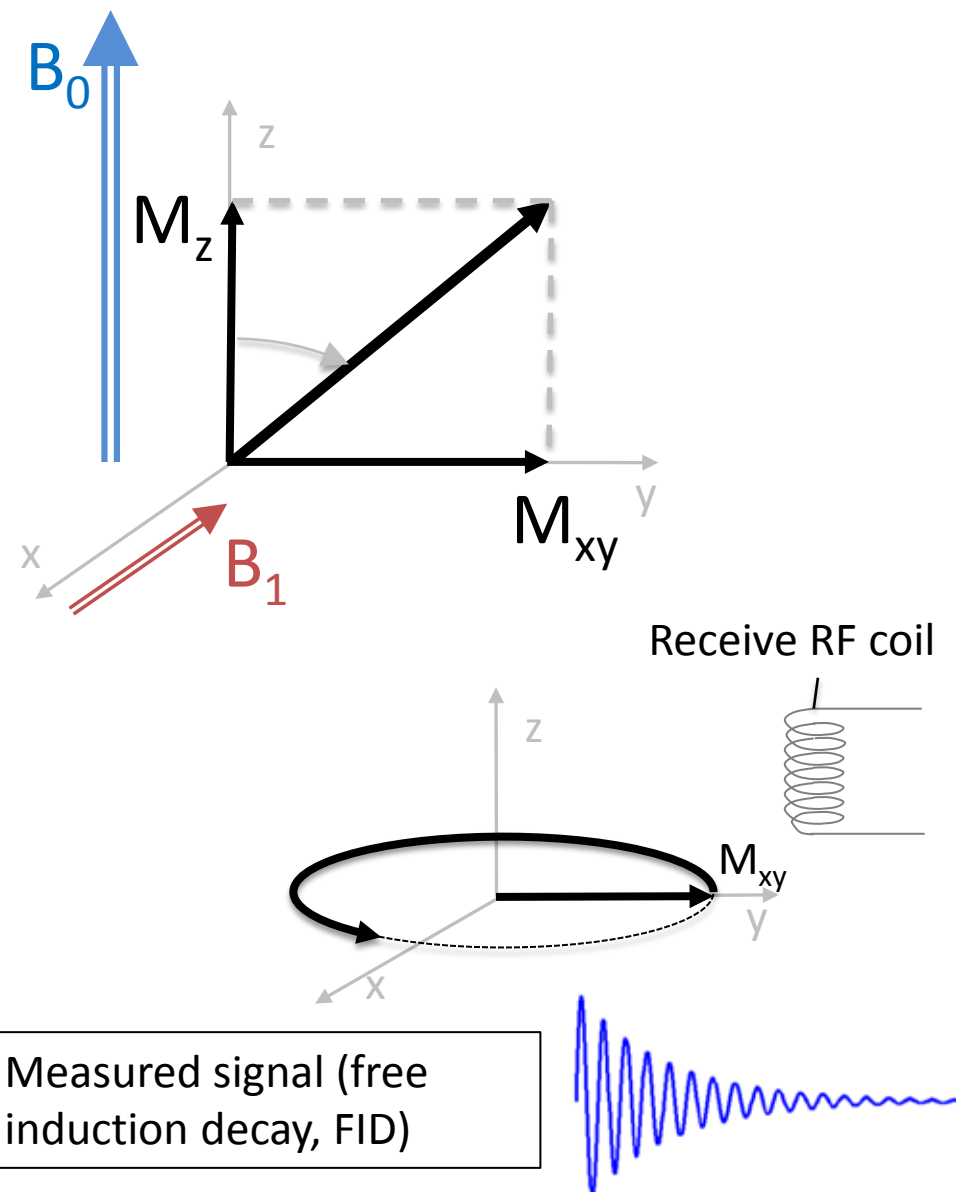
Transverse Decay - T_2 /T_2*

- Differences in the resonant frequencies of individual spins cause dephasing over time reducing the transverse magnetisation M_{xy}.

- The frequency differences are due to the influence of small magnetic fields around neighbouring spins affecting the magnetic field an individual spin experiences.

- The rate of dephasing is described by the decay constant T_2.

- In practice the rate of dephasing is further increased by inhomogeneities in B_0, with the cumulative decay rate described by the decay constant T_2* which is shorter than T_2.

Spin dephasing

M_{xy}

M_o

$----$ No B_0 inhomogeneities : $M_{xy} = M_0 e^{-t/T_2}$

$-----$ With B_0 inhomogeneities: $M_{xy} = M_0 e^{-t/T_2*}$

t

Longitudinal Recovery – T_1

- After the RF pulse the spins gradually return to their equilibrium state with a net magnetisation M_0 parallel to B_0.

- The recovery rate is described by the constant T_1.

- A long T_1 gives slow recovery whereas a short T_1 gives fast recovery.

- T_1 is always longer than T_2.

$$M_z = M_0 \left(1 - e^{-t/T_1} \right)$$
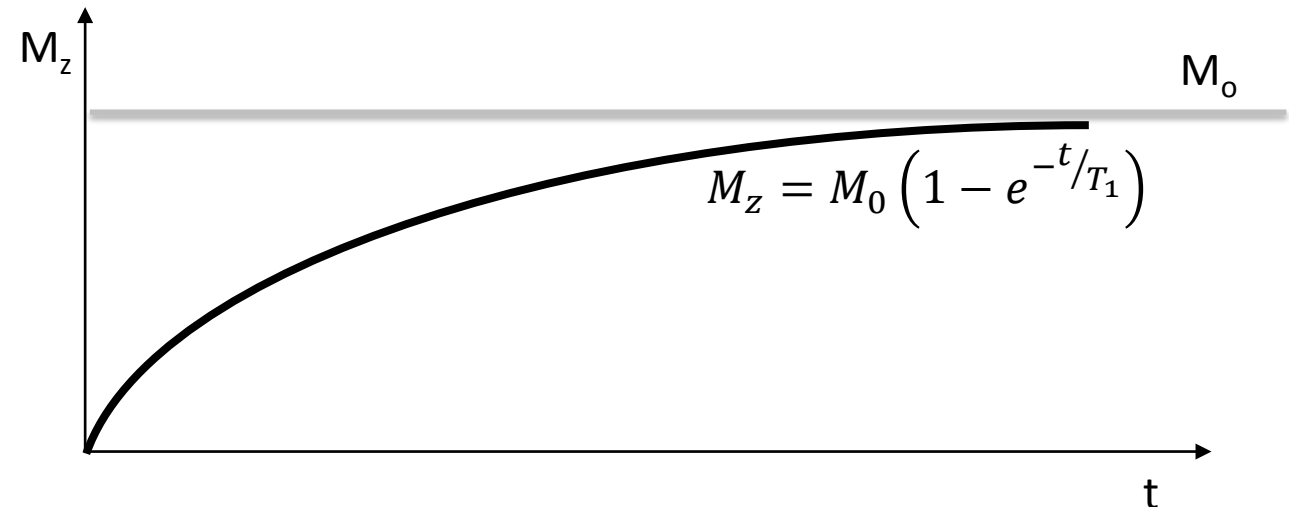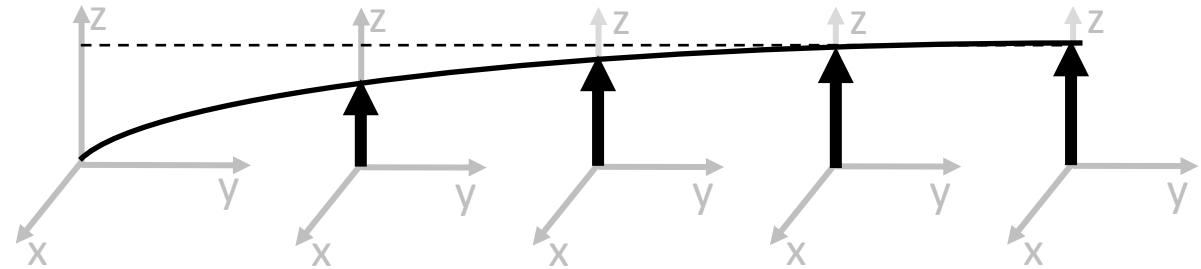
Background

- Signal intensity is determined by the relaxation times of the tissues being imaged and by the chosen scan parameters. Contrast agents can be administered, to alter the T_1 and T_2 properties of tissues and so alter image contrast.

- The most commonly used agents are gadolinium based contrast agents (GBCAs, a chelate molecule is required to make the agent non-toxic).

- These are injected intravenously and are distributed throughout the body. The most commonly used agents (extracellular agents) leak out of capillaries into interstitial spaces, but do not cross intact cell membranes.

- The agents are (predominantly) cleared by renal excretion.

Physical Effects

- The main effect of GBCAs is to shorten the T_1 of the surrounding tissue. They also exhibit a T_2/T_2^* shortening effect although this is a much smaller effect for the concentration in tissue achieved in most clinical contrast doses. GBCAs are therefore used to increase signal in T_1 weighted images.

Relaxivity

- GBCAs lead to an increase in relaxation rates proportional to relaxivity and concentration of the agent:

$$R_1 = R_1(\text{native}) + [\text{GBCA}] \times r_1$$
$$R_2 = R_2(\text{native}) + [\text{GBCA}] \times r_2$$

Where R_1 and R_2 are relaxation rates ($R_1 = 1/T_1$ and $R_2 = 1/T_2$), r_1 and r_2 are longitudinal and transverse relaxivities, [GBCA] is contrast agent concentration

Slice Selection

- RF excitation only occurs when the RF frequency matches the resonant frequency of the spin.

- A magnetic field gradient G_{SS} is applied during the RF pulse such that the resonant frequencies of the spins vary spatially.

- By limiting the bandwidth of frequencies in the RF transmit pulse, only spins within a given slice will be excited.
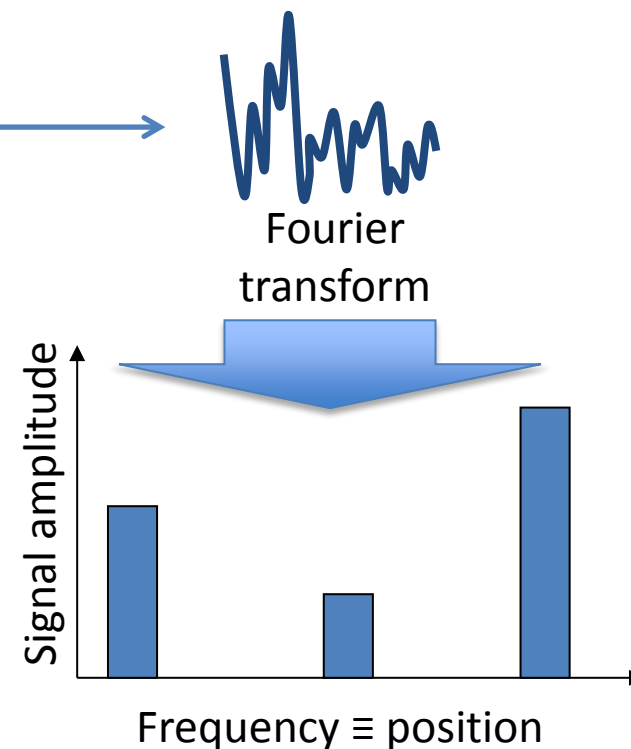
Magnetic field strength, B

G_{SS}

B_0

Excited slice

Resonant frequency, ω

RF bandwidth
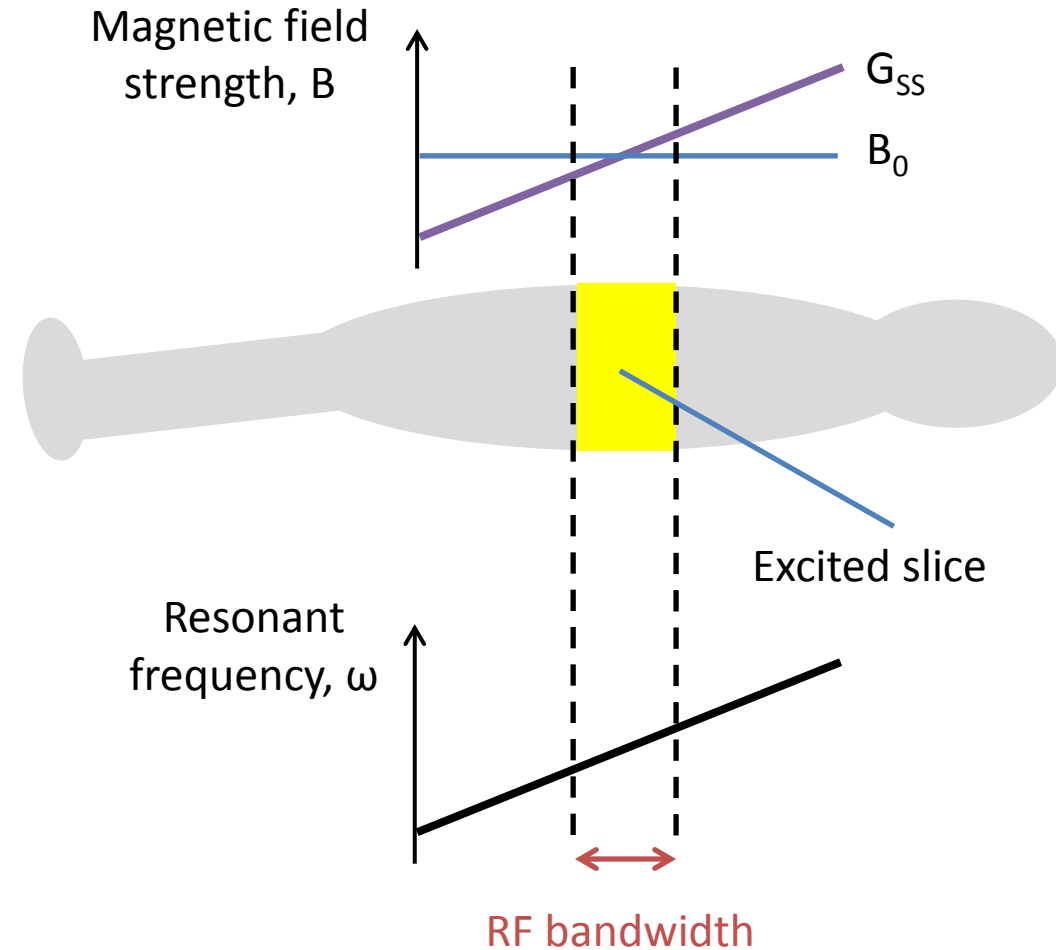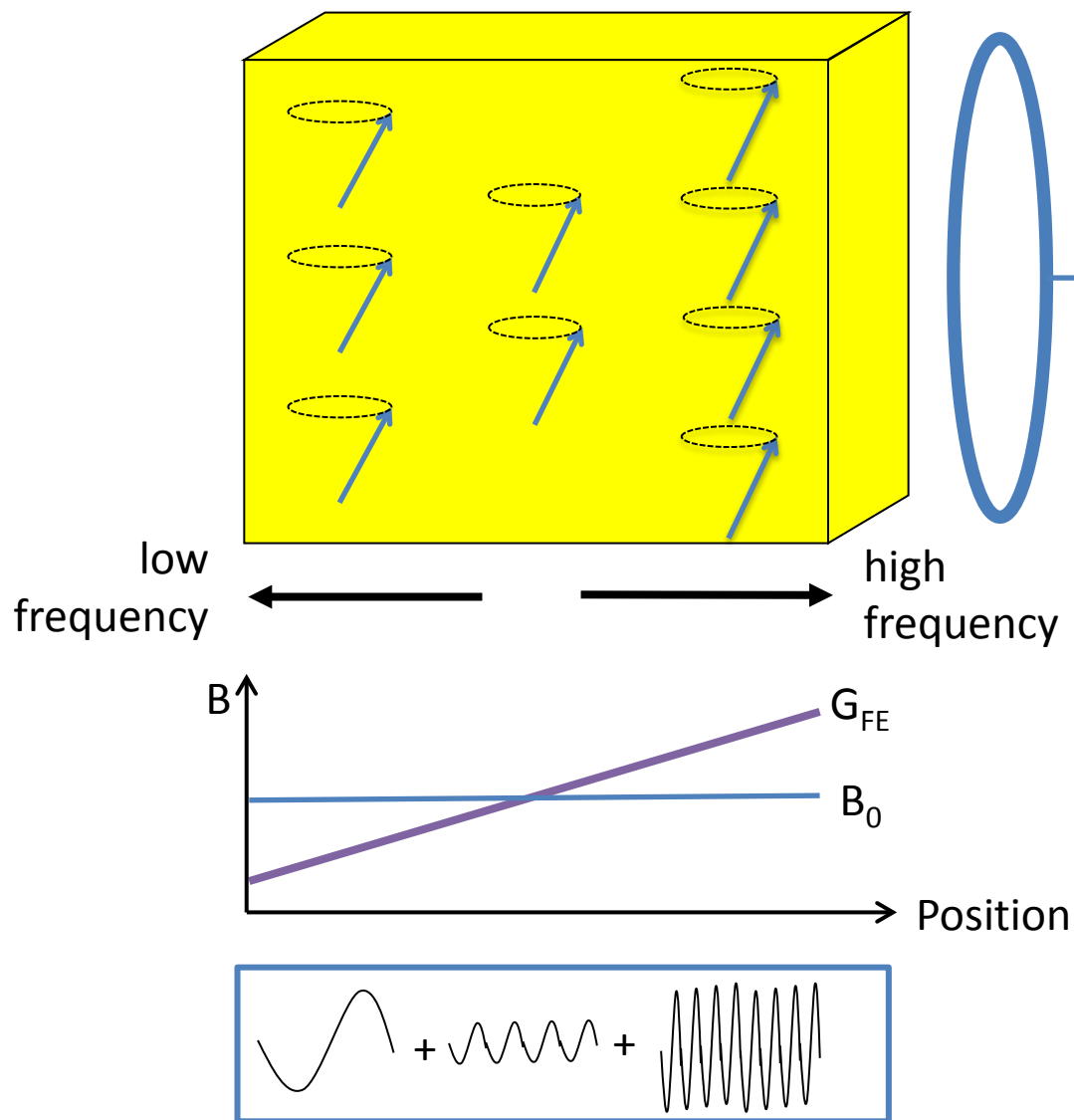
Frequency Encoding

The received signal is the sum of the frequencies from all the spins

low frequency

high frequency

RF receiver coil

Fourier transform

B

G_{FE}

B_0

Position

Signal amplitude
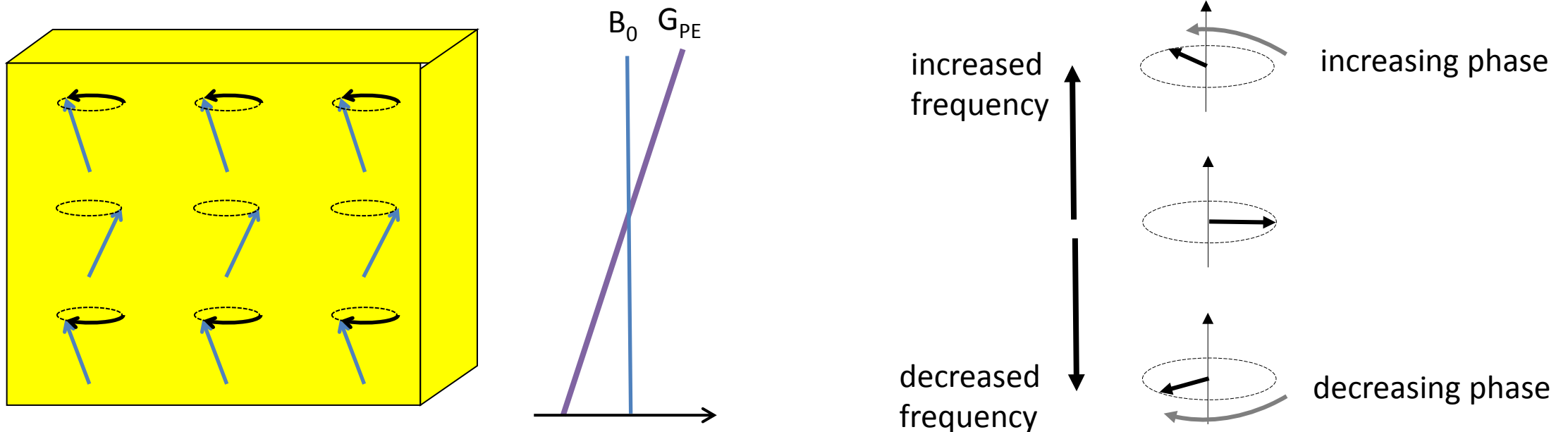
Frequency ≡ position

The result of the Fourier transform plots the signal amplitude at each frequency component, corresponding to location along the frequency encoding direction

Frequency Encoding

- Frequency encoding uses magnetic field gradients to vary the resonant frequencies of spins across space so that the position of a signal can be obtained from its spatial frequency.

- A magnetic field gradient G_{FE} is applied during data acquisition such that the resonant frequencies of the spins vary spatially. The position of a given spin is now proportional to its frequency.

- The measured signal is the sum of all of the frequencies within the image slice.

- A Fourier transform is used to convert this summed signal from the frequency domain to the spatial domain. Because position along the frequency encoding gradient is proportional to frequency, the Fourier transform arranges the data as it was originally positioned in the slice along the frequency encoding direction.
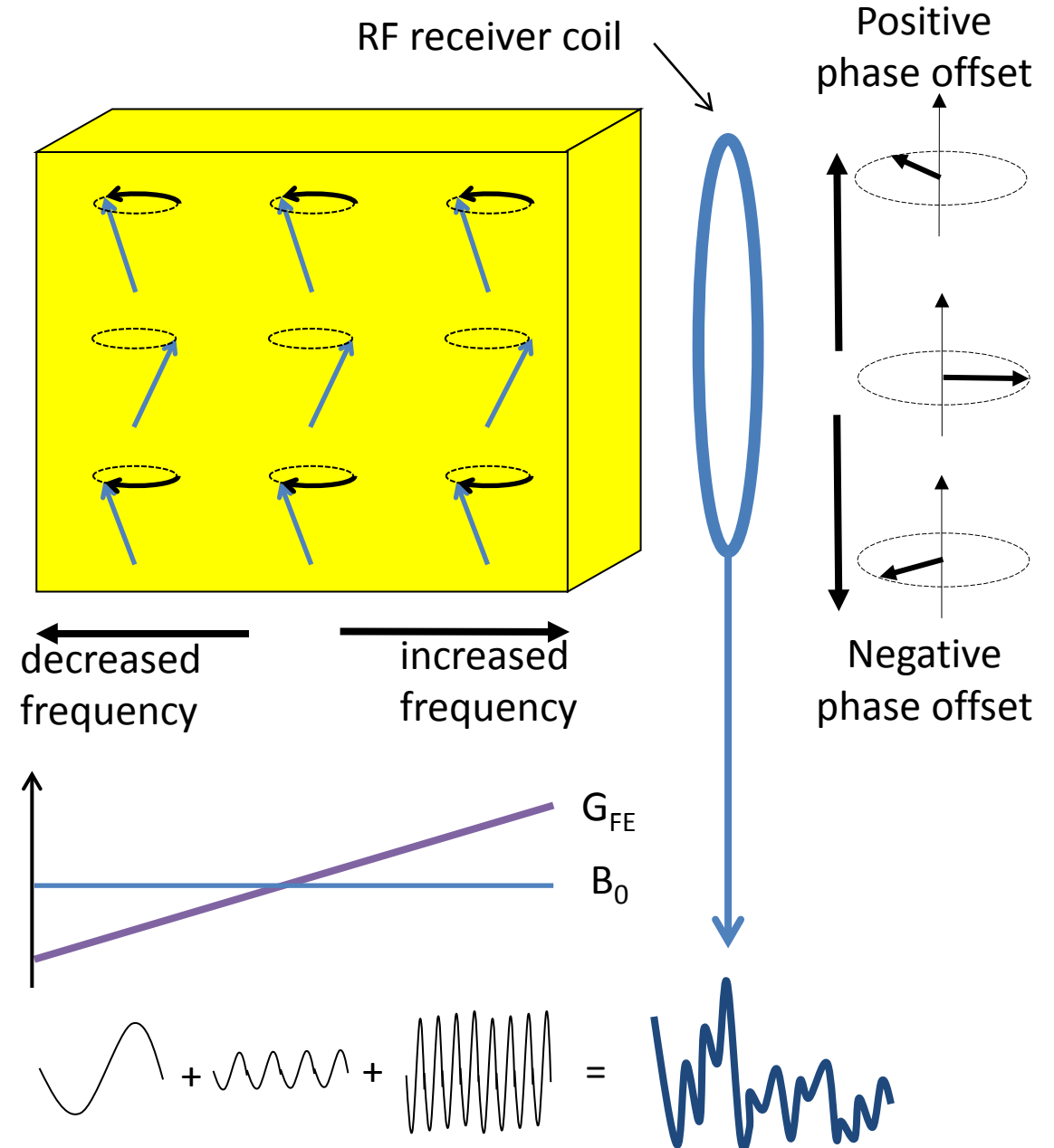
Phase Encoding

- Prior to frequency encoding, a phase encoding gradient G_{PE} is applied along the image plane at right angles to G_{PE}.

- Whilst G_{PE} is applied the frequency in the phase encoding direction is proportional to the position along G_{PE}.

- After G_{PE} is turned off each spin has acquired a phase offset proportional to the change in frequency whilst the phase encoding gradient was on.

- This phase offset is now a function of position in the phase encoding direction.
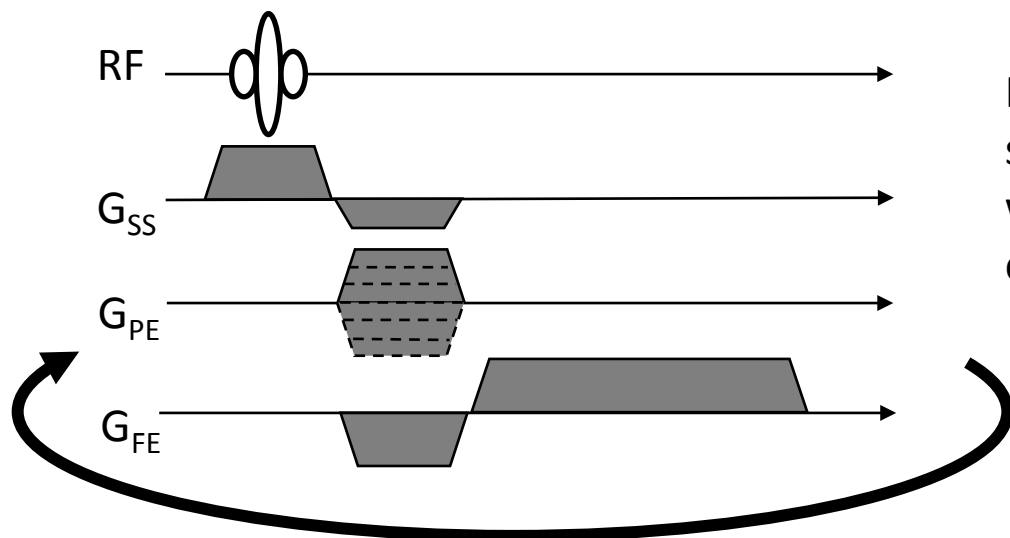
Phase Encoding

- After G_{PE} is turned off the frequency encoding read-out gradient is applied.

- The spins, which still have a phase offset in phase encoding direction now have a frequency dependent on their position in the frequency encoding direction.

- The MR signal is measured whilst the frequency encoding gradient is switched on.

- Within the measured signal frequency encodes position in the G_{FE} direction and phase encodes position in the G_{PE} direction.

RF receiver coil

Positive phase offset

decreased frequency   increased frequency

Negative phase offset

G_{FE}

B_0

K-space

RF
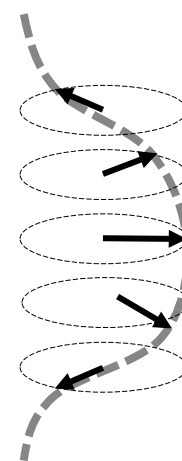
G_{SS}

G_{PE}

G_{FE}

TR

Pulse sequence acquires signal multiple times with different phase encoding gradients
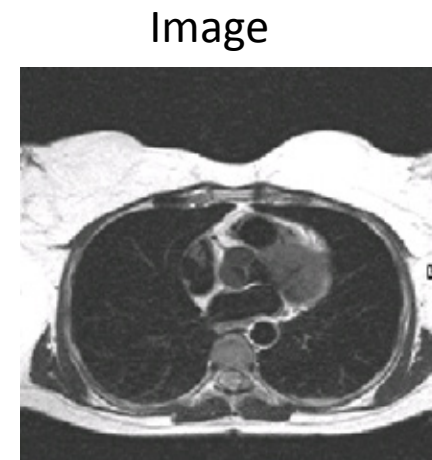
Phase changes over time appear as an oscillating signal. The rate of change of phase can then be Fourier transformed into the spatial domain.

K-space

Image
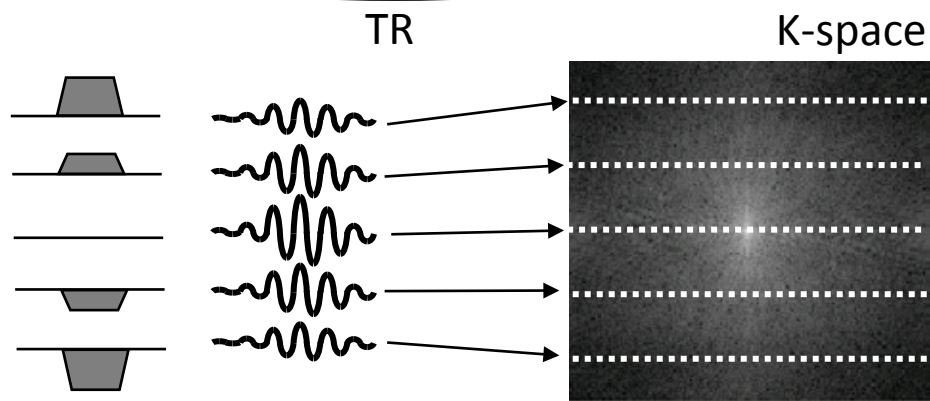
Fourier Transform

Phase encoding gradient determines which line of k-space is filled

K-space

- K-space is the name of the data array within which the multiple signal measurements that make up a single MR image is saved.

- The Fourier transform requires time varying data, so a single phase encoding step is inadequate as it only generates a signal with a single phase offset. The MR signal must therefore be measured repeatedly with multiple phase encoding steps.

- Successive phase encoding steps alter the gradient strength in equal increments. This gives rise to equal increments in the phase offset. The size of each increment relates to the position along the phase encoding gradient.

- In conventional imaging the time interval between each acquisition is TR.

- The two dimensional Fourier transform of k-space transforms the data from the time domain into the frequency domain. Because frequency and rate of change of phase are proportional to position along the respective gradient directions, the locations in the frequency domain following the Fourier transform will correctly represent the spatial locations of the signal components in the final image.

- Instead of imaging multiple slices separately it is possible to perform 3D imaging in which a slab of tissue is excited and localisation is performed by frequency encoding in one dimension and by phase encoding and in the other two.
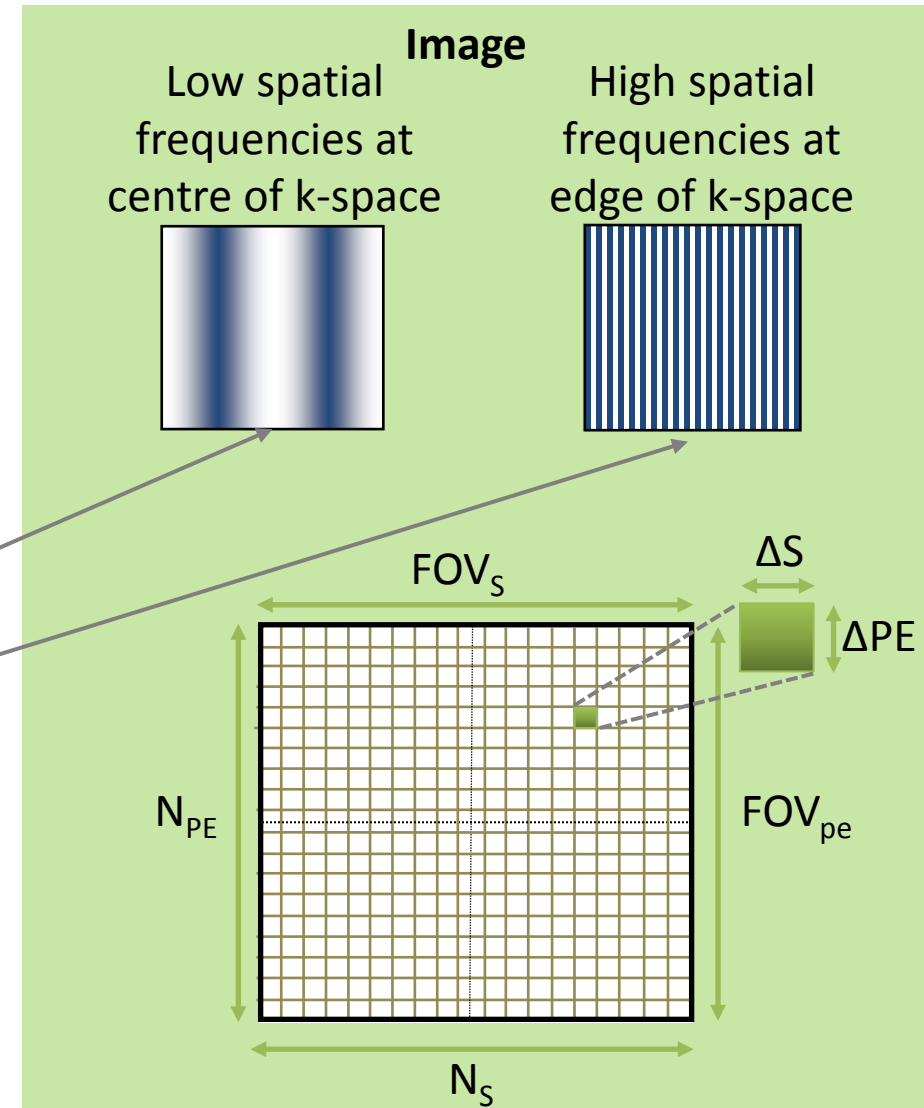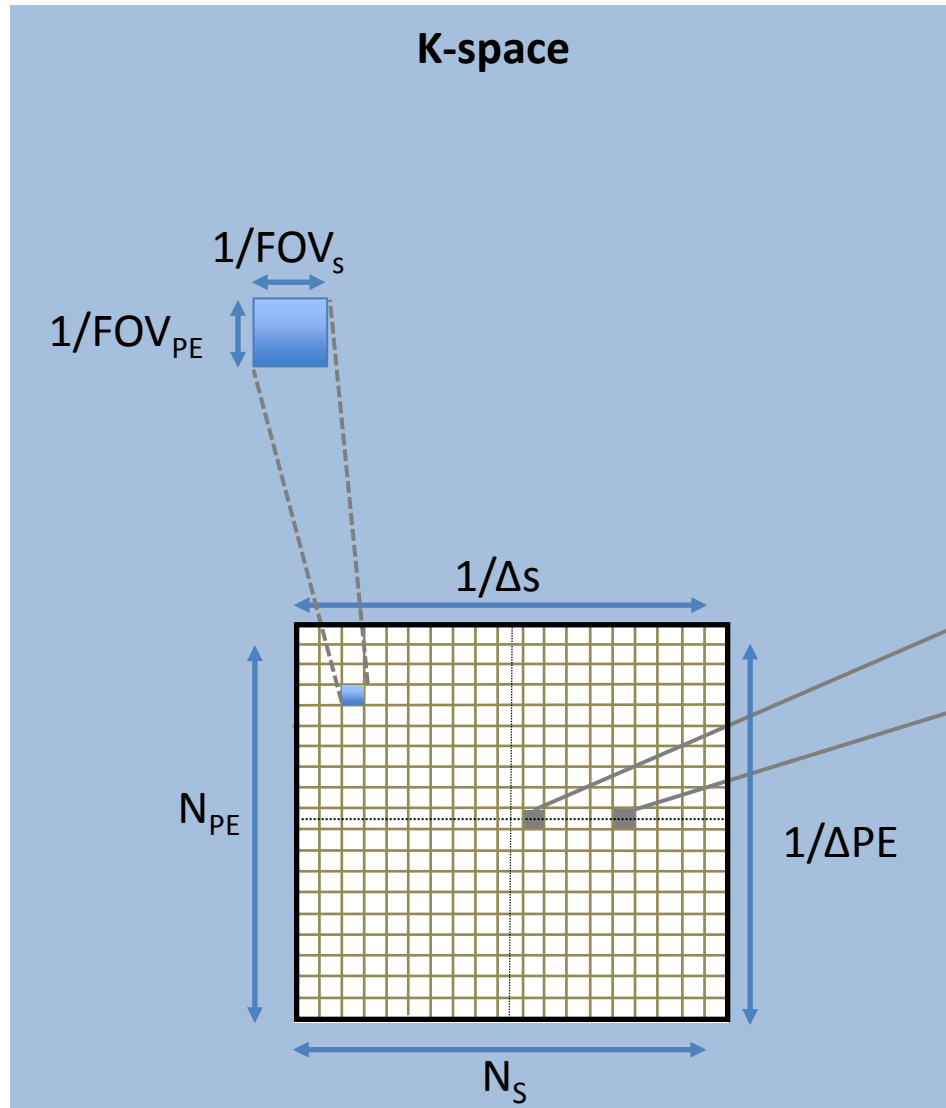
K-space Properties

- Each point in k-space represents a single spatial frequency's contribution across the whole image. Similarly every MR voxel signal intensity is influenced by every spatial frequency in k-space.

- Low spatial frequency contributions are found at the centre of k-space and high spatial frequencies at the edges.

- The number of data points in k-space corresponds to the number of voxels in the image. The number of samples N_S that each signal is digitised into is the number of voxels in the frequency encoding direction. In conventional imaging the number of phase encoding steps, N_{PE}, is the number of voxels in the phase encoding direction.

- The field of view (FOV) of the image is inversely related to the spacing between data points in k-space in each direction.

- The voxel size in the image is determined by the dimensions of k-space i.e. how far the data points lie from the centre of k-space.

SNR

- Every MR voxel contains a mixture of signal and noise. The MR signal is the electrical voltage induced in the receiver coil by the precession of the transverse component of the net magnetization M_{xy} around the longitudinal axis. The noise is due to unwanted random electrical fluctuations from other sources. These exist in all conducting materials, including MR receiver coils and human tissue.

- The signal to noise ratio (SNR) is the ratio of these two contributions :

$$SNR = \frac{signal}{noise}$$

- SNR increases with voxel volume and the number of signal samples (i.e. the number of data points in k-space) but decreases with increasing receiver bandwidth (rBW):

$$SNR \propto \frac{(voxel\ volume)\sqrt{No.\ of\ signal\ samples}}{\sqrt{rBW}}$$

- The number of signal samples is proportional to the number of phase encoding steps.

Acquisition Time

- In conventional imaging the image acquisition time is proportional to the repetition time and the number of lines of k-space acquired (i.e. the number of phase encoding steps):

$$Image\ Acquisition\ Time = TR \times N_{PE}$$

- An increase in image resolution in the phase encoding direction incurs an time penalty as the increase in the number of phase encoding steps requires additional repetitions in conventional imaging pulse sequences.

- Frequency encoding does not incur a similar penalty as this only increases the data acquisition time for each signal echo, without changing the number of repetitions or necessarily affecting the repetition time.

Receiver Bandwidth (rBW)

- The receiver bandwidth is the range of frequencies that will be received by the imaging system.

- The receiver bandwidth is set by a combination of the frequency encoding gradient strength and the FOV. For a given FOV, a larger gradient will create a higher maximum resonant frequency (f_{max}) and increase the range of frequencies between f_{max} and f_{min}, hence a larger frequency bandwidth.

- The measured signal must be sampled at a sufficient sampling frequency to represent f_{max} in k-space. The sampling frequency must be at least twice f_{max} to avoid signal aliasing, so $f_s = 2f_{max}$.

- The sampling interval T_s is the time between adjacent sampling points, and is the reciprocal of f_s i.e. $T_s = 1/f_s$.

- The total sampling time, T, is the sum of all the sampling intervals and limits the minimum possible echo time (TE_{min}).

- Electronic noise is distributed evenly across frequencies. Thus high rBW values have poorer SNR than low rBW values because more noise is included across a wider frequency range. However, high rBW values allow a shorter TE and can minimise some artefacts (e.g. chemical shift, geometric distortion).

Receiver Bandwidth (rBW)

Frequency encoding gradient

f_{max}

B_0

rBW

f_{min}

Measured signal

N_s

T_s

T
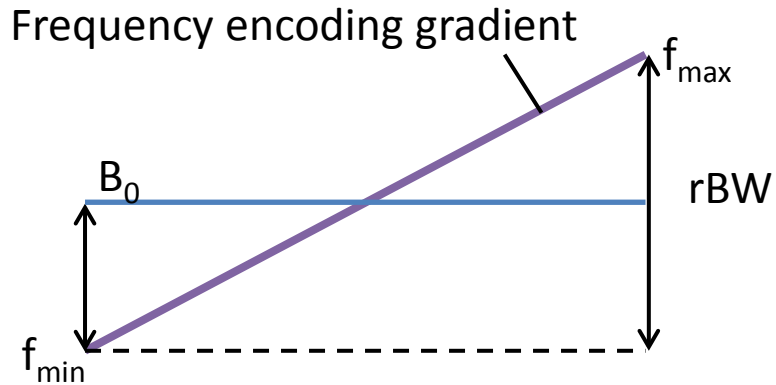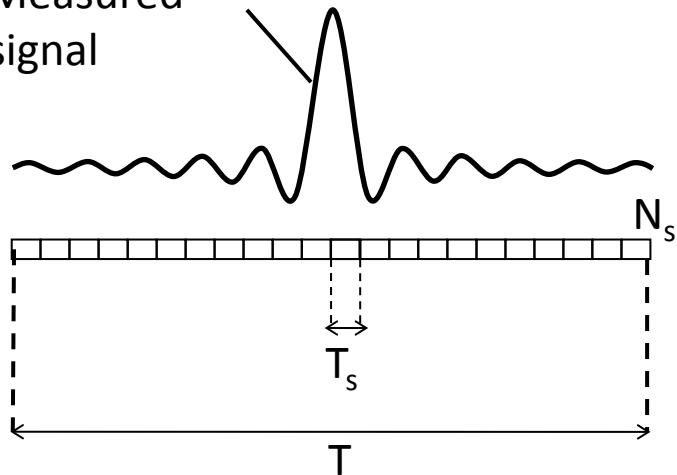
Low rBW (low rBW leads to longer T)

Frequency encoding gradient

f_{max}

rBW

f_{min}

Measured signal

N_s

T_s

T
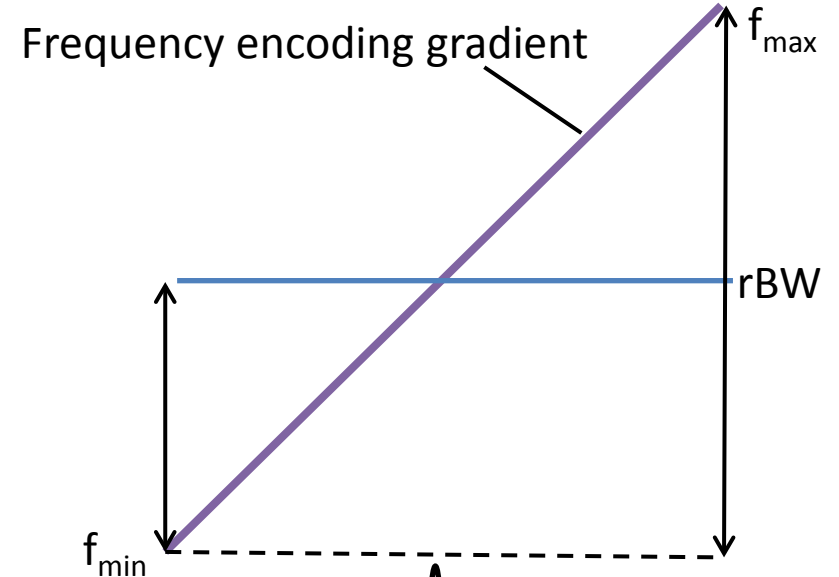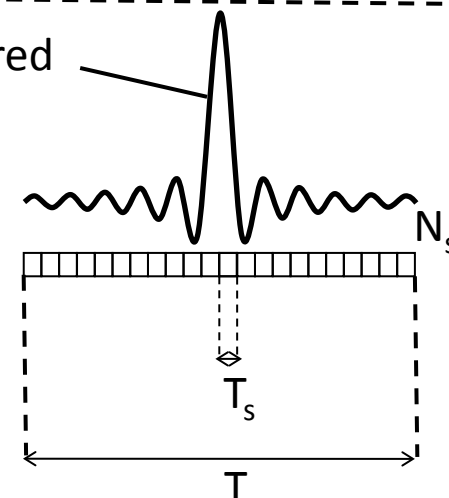
High rBW (high rBW leads to shorter T)

Full Sampling

- Acquiring the full k-space data set can be time consuming (equal to the TR multiplied by N_{PE}). Acquisition times can be reduced by decreasing the number of lines of k-space that are acquired. Different strategies are used to under-sample k-space, each with associated trade-offs.
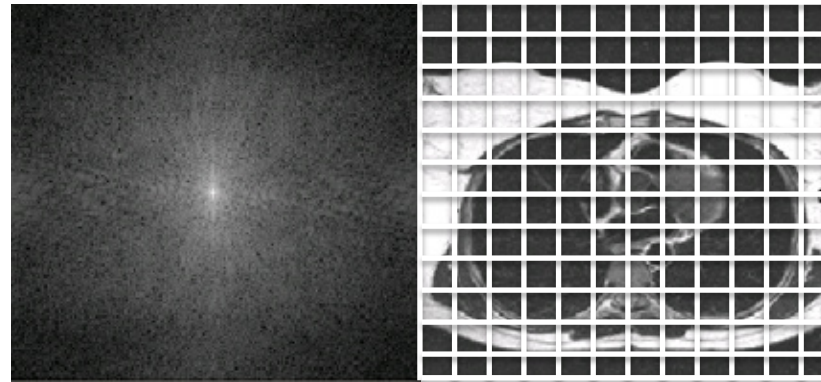
$$SNR \propto \frac{(voxel\ volume)\sqrt{No.\ of\ signal\ samples}}{\sqrt{rBW}}$$

Under-Sampling k-space – Partial Fourier

- Partial Fourier imaging acquires just over half of the k-space and then exploits the complex conjugate symmetry of k-space to generate the unsampled lines. The final k-space has all N_{PE} lines of k-space so the image resolution is maintained. However, less lines of k-space have been physically acquired so the SNR will be reduced.
- Zero filling, where the missing lines of k-space are assigned zero values, is often used as an alternative to complex conjugate synthesis.
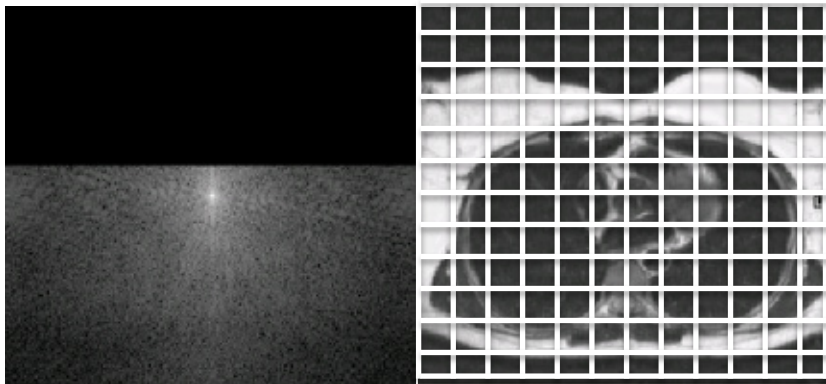
Resolution	Unchanged- k-space dimensions unaltered
SNR	Reduced- less lines acquired
Acquisition time	Reduced- less lines acquired

Under-Sampling k-space – Reduced Acquisition Matrix

- Reducing the acquisition matrix maintains the distance between k-space lines, so that k-space does not extend out to higher frequencies in the phase encoding direction. Although the number of lines of k-space are reduced the voxel size is increased in the phase encoding direction resulting in a net increase in SNR and decreased resolution.
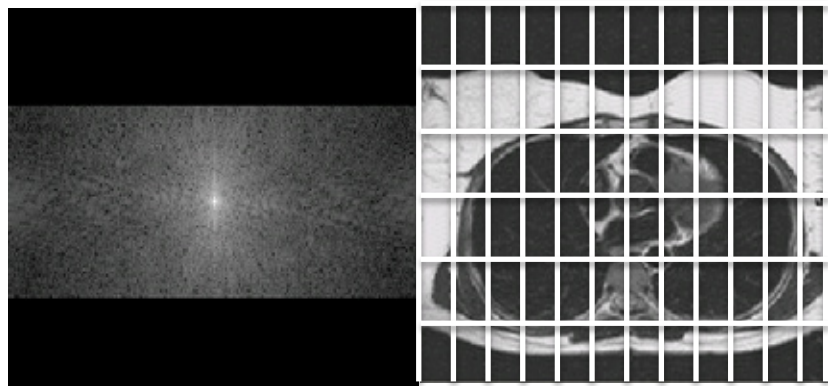
Resolution	Reduced- less lines acquired
SNR	Increased- larger voxel size
Acquisition time	Reduced- less lines acquired

Under-Sampling k-space – Rectangular Field Of View

- In imaging with a rectangular field of view (rFOV) data is acquired at the same extent in k-space but with increased spacing between the lines, resulting in a reduced FOV in the phase encoding direction. Pixel size (spatial resolution) remains the same but SNR is reduced as fewer lines of data are sampled.

- If the FOV does not extend over all of the signal generating object then this will result in image aliasing or wrap-around artefact. Because the spatial encoding gradients continue outside of the FOV magnetic moments outside the FOV will acquire a rate of change of phase offset of greater than 180°, (or less than -180°). Because a rate of change of phase of, say, 190° is indistinguishable from -170° these signals will be mapped to the -170° position, appearing on the opposite side of the image.
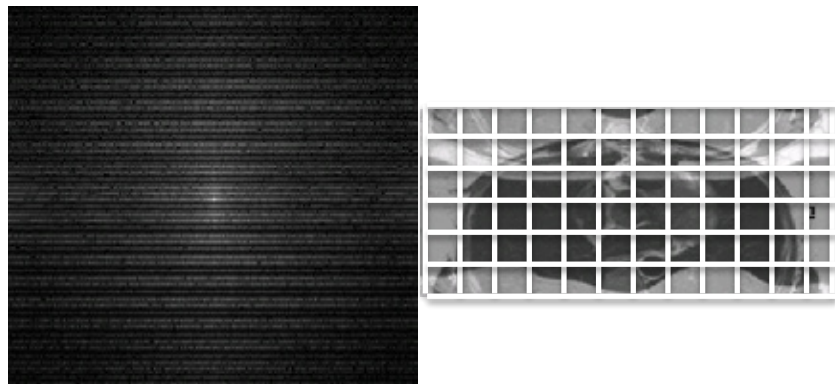
Resolution	Unchanged- full extent of k-space sampled.
SNR	Reduced- less lines acquired
Acquisition time	Reduced- less lines acquired

Parallel Imaging

- Parallel imaging uses the spatial distribution of multiple coils or coil elements and their characteristic sensitivity maps to provide spatial information, which allows under-sampling of k-space, shortening the acquisition time.

- The image is acquired using a reduced number of k-space lines. The same extent of k-space is covered but fewer lines are acquired so that the resolution is maintained but the FOV is reduced causing image aliasing. The factor by which the number of lines is reduced is called the reduction factor, R.

- A reference image in which only the central lines of k-space are acquired is also used. The resulting low resolution, full FOV image is used to measure the coil sensitivity profiles for each individual coil.

- By considering the aliased image generated by each separate coil along with extra information provided by the coil sensitivity profile it is possible to write multiple independent mathematical expressions for the true voxel intensity. These equations can be solved to generate the final, un-aliased image.

- The reference image is either acquired prior to parallel imaging (SENSE, ASSET) or as part of the acquisition (mSENSE, GRAPPA, ARC). The reconstruction calculations can be performed in image space (SENSE, ASSET) or in k-space (SMASH, GRAPPA, ARC).
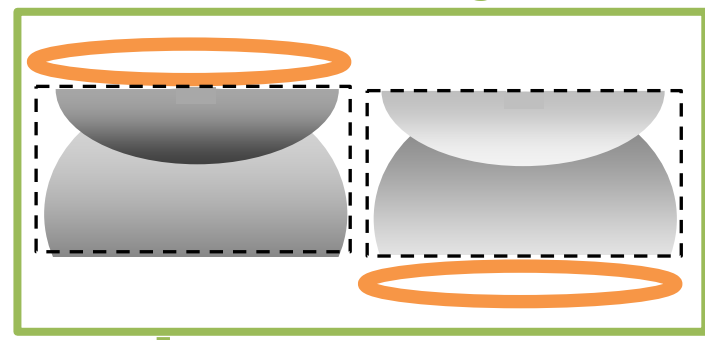
Parallel Imaging

Coil 1

FOV

Object

Coil 2
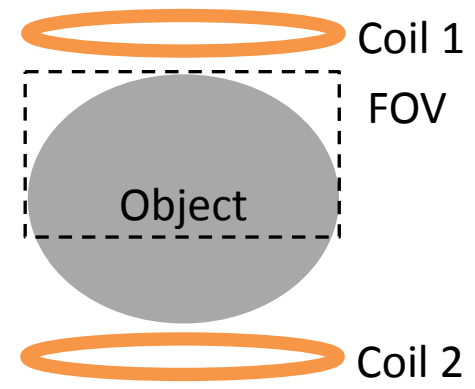
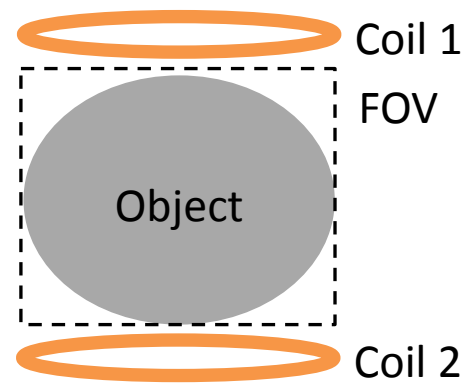R factor of 2 skips alternate lines

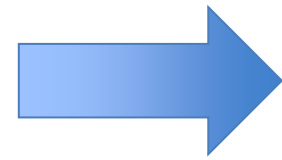Aliased images

Coil 1

FOV

Object

Coil 2
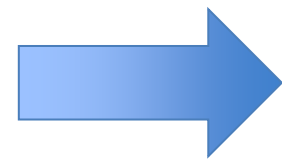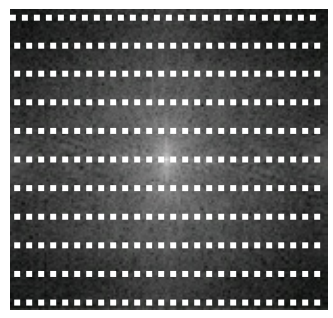
Full FOV low resolution image

Fourier transform

RECONSTRUCTION ALGORITHM

Final image
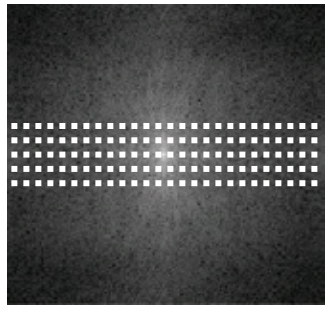
Coil sensitivity maps
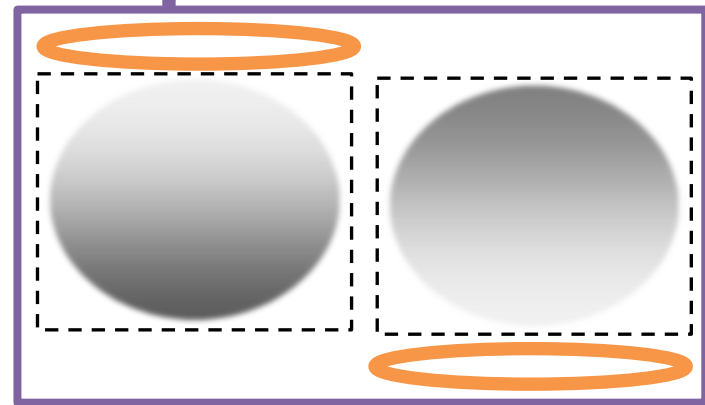
Source of Image Contrast

- The signal intensity at a particular voxel is dependent on:
    1. The coil sensitivity at that spatial location.
    2. Scanner hardware parameters.
    3. The equilibrium magnetisation within that voxel (increases with field strength and proton density).
    4. The magnitude of the transverse magnetisation at the time of signal data acquisition:
        - This is dependent on tissue characteristics (e.g. T_1, T_2 and T_2*) and scan parameters (e.g. TR, TE, RF pulse flip angles) and the pulse sequence
    5. Other effects (e.g. flow, diffusion, signal averaging).

Contrast weighting

- The contrast weighting sets the extent to which a given tissue characteristic effects the image contrast.

- By varying scan parameters the dependence of the signal contrast on tissue characteristics such as the relaxation time can be altered.

- Contrast weighting is achieved by selecting sequence parameters that reduce signal intensity from the maximum possible value. Consequently signal to noise ratio (SNR) will be lower in heavily weighted sequences than in proton density weighted sequences. Selection of appropriate parameters will depend on a compromise between desired weighting and SNR.

Impact of TE

- The Echo Time (TE) defines the amount of time between the excitation pulse and the signal data acquisition. The longer the TE, the lower the signal from all tissues, but signal decreases more rapidly for tissues with short T_2 (spin-echo) or T_2^* (gradient echo).

T_2/T_2^* Weighting and SNR

- Short TE minimises differences in signal due to transverse magnetisation decay and maintains high signal and SNR.

- Longer TE increases relative differences in signal due to transverse decay increasing $T_{2/}T_2^*$ weighting.

- As TE increases signal, and so SNR, decreases. TE must be optimised for both weighting and SNR.
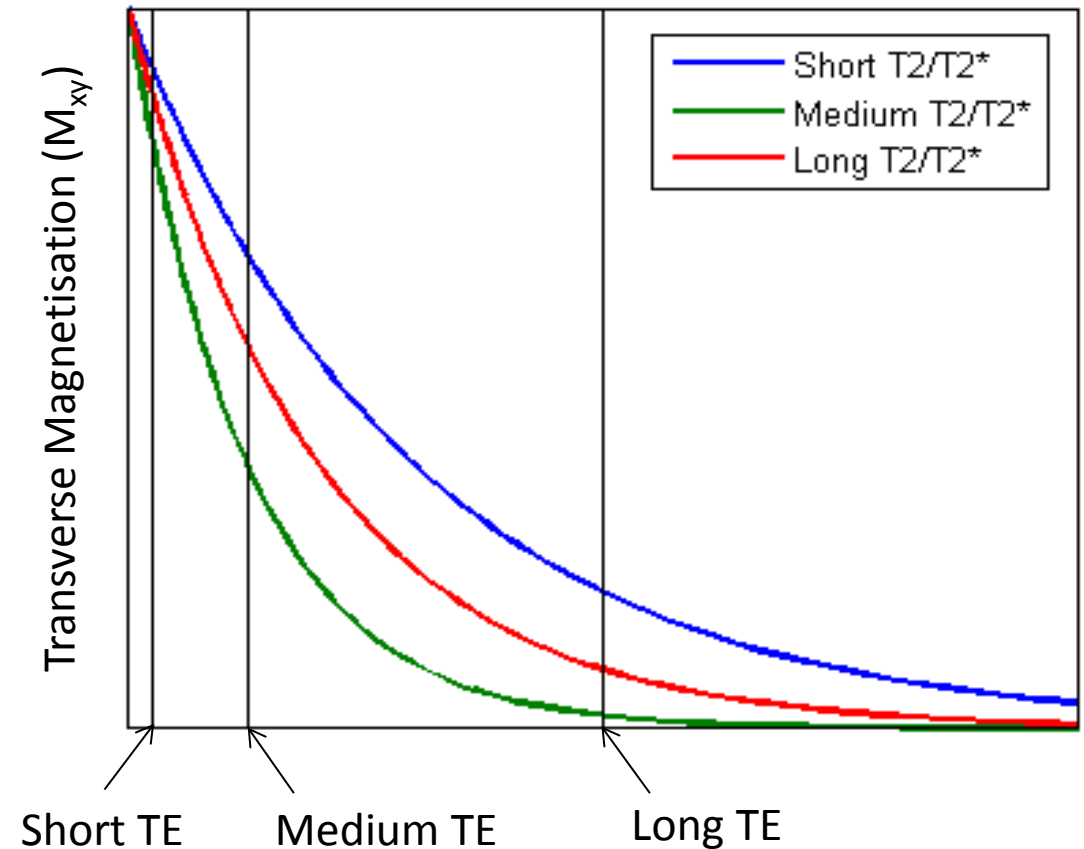
Impact of TR

- The Repetition Time (TR) defines the amount of time between excitation pulses. Longer TR allows more complete longitudinal magnetisation recovery and so higher signal from all tissues, but signal recovers more rapidly for tissues with short T_1.

T_1 Weighting and SNR

- Long TR allows near complete recovery for all tissues, so differences due to T_1 recovery are small and SNR is high.

- Shorter TR increases relative differences in signal due to T_1 recovery so increases T_1 weighting.

- As TR decreases signal, and so SNR, also decreases. TR must be optimised for both weighting and SNR.
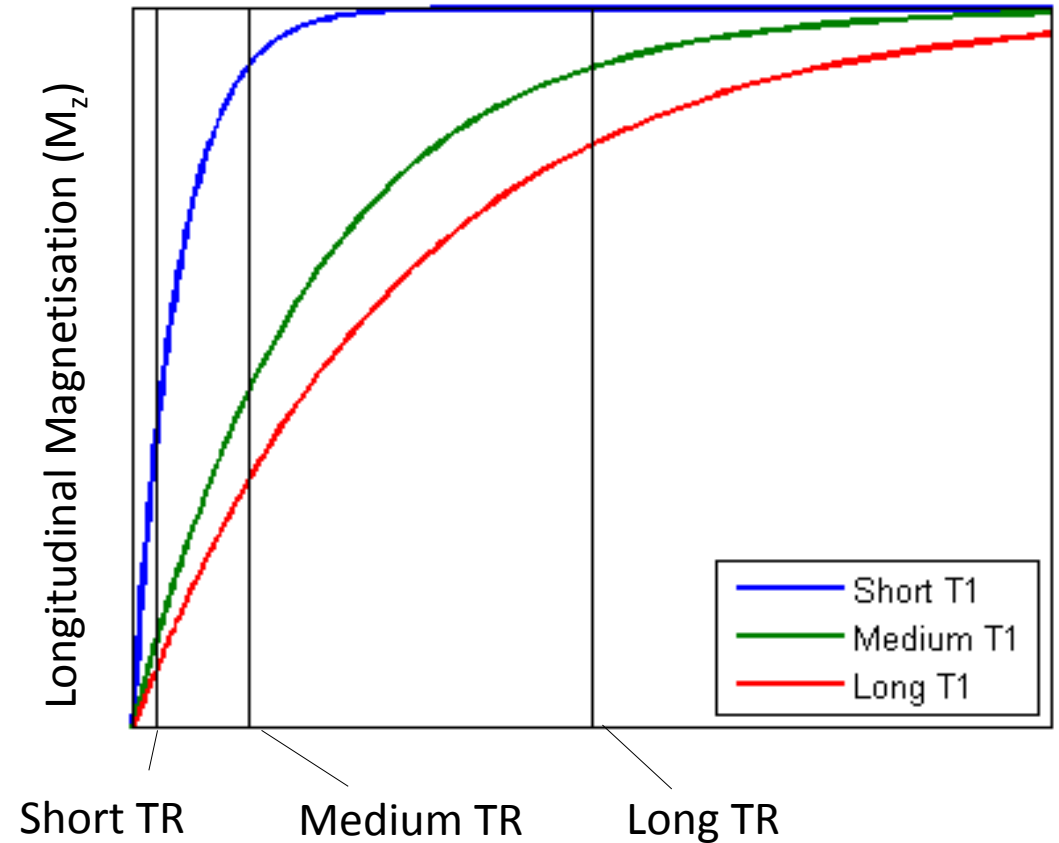
Mechanism

- T_1 weighting is increased by choosing scan parameters that enhance the relative differences in the longitudinal magnetisation at the time the readout pulse is applied.

Magnetisation Evolution

- Short TR enhances differences in longitudinal magnetisation.

- Short TE minimises differences due to transverse decay.

- Signal differences mostly due to differences in T_1.

Longitudinal Magnetisation

TR

M_z

Fat - Short T_1

RF pulse

Soft Tissue - Moderate T_1

Fluid - Long T_1

Transverse Magnetisation

M_{xy} TE

M_{xy}

Fat - Moderate T_2

Fluid - Long T_2

Spoiler

Soft Tissue - Short T_2
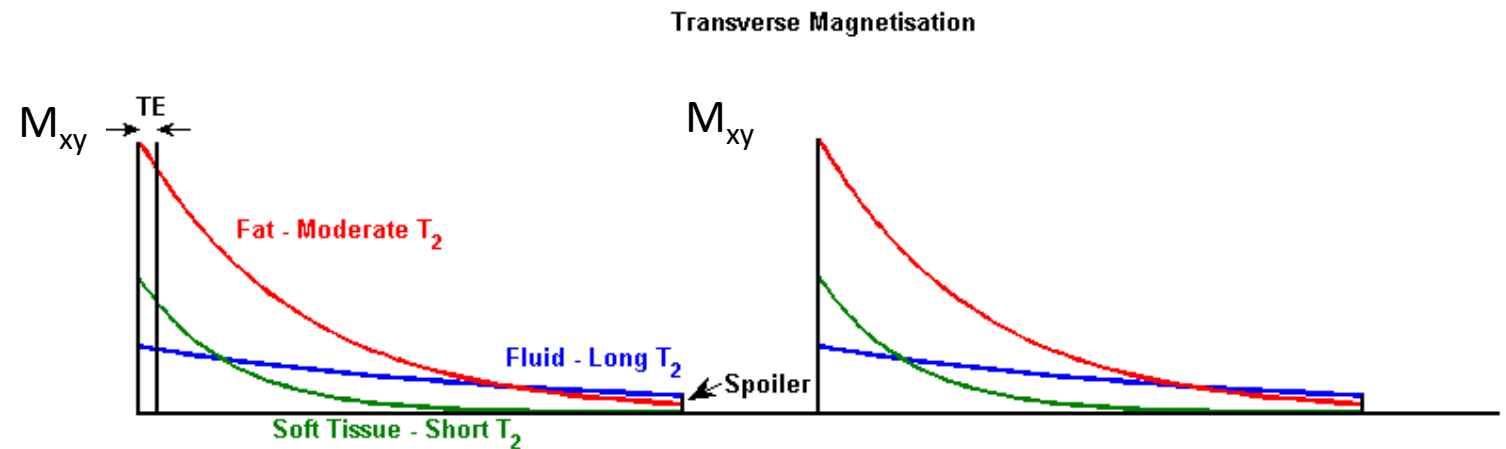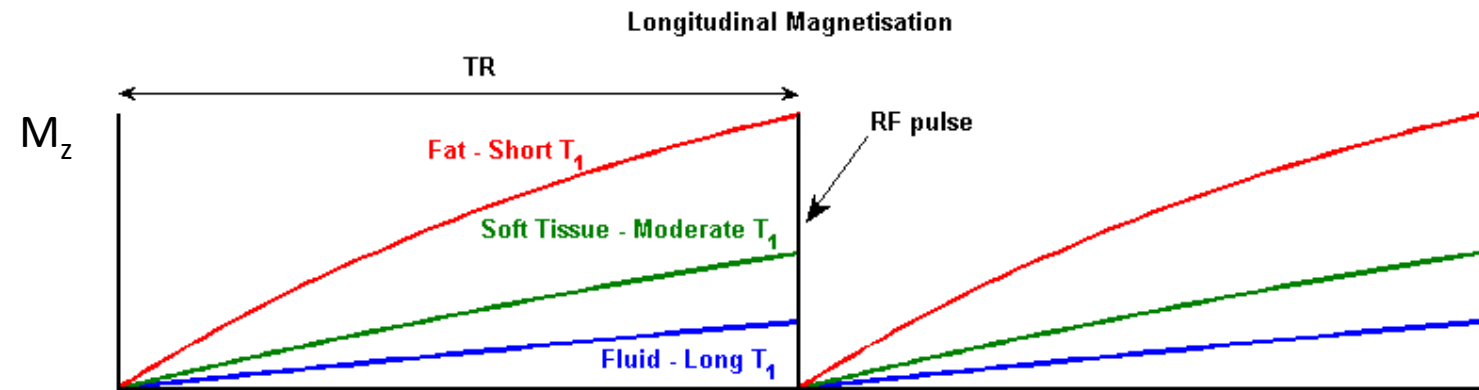
Mechanism

- T$_2$ or T$_2$* weighting is increased by choosing scan parameters that enhance the relative differences in the transverse magnetisation at the time the signal is received.

Magnetisation Evolution

- Long TR minimises differences in longitudinal magnetisation.

- Long TE introduces differences due to transverse decay.

- Signal differences mostly due to differences in T$_2$ or T$_2$*
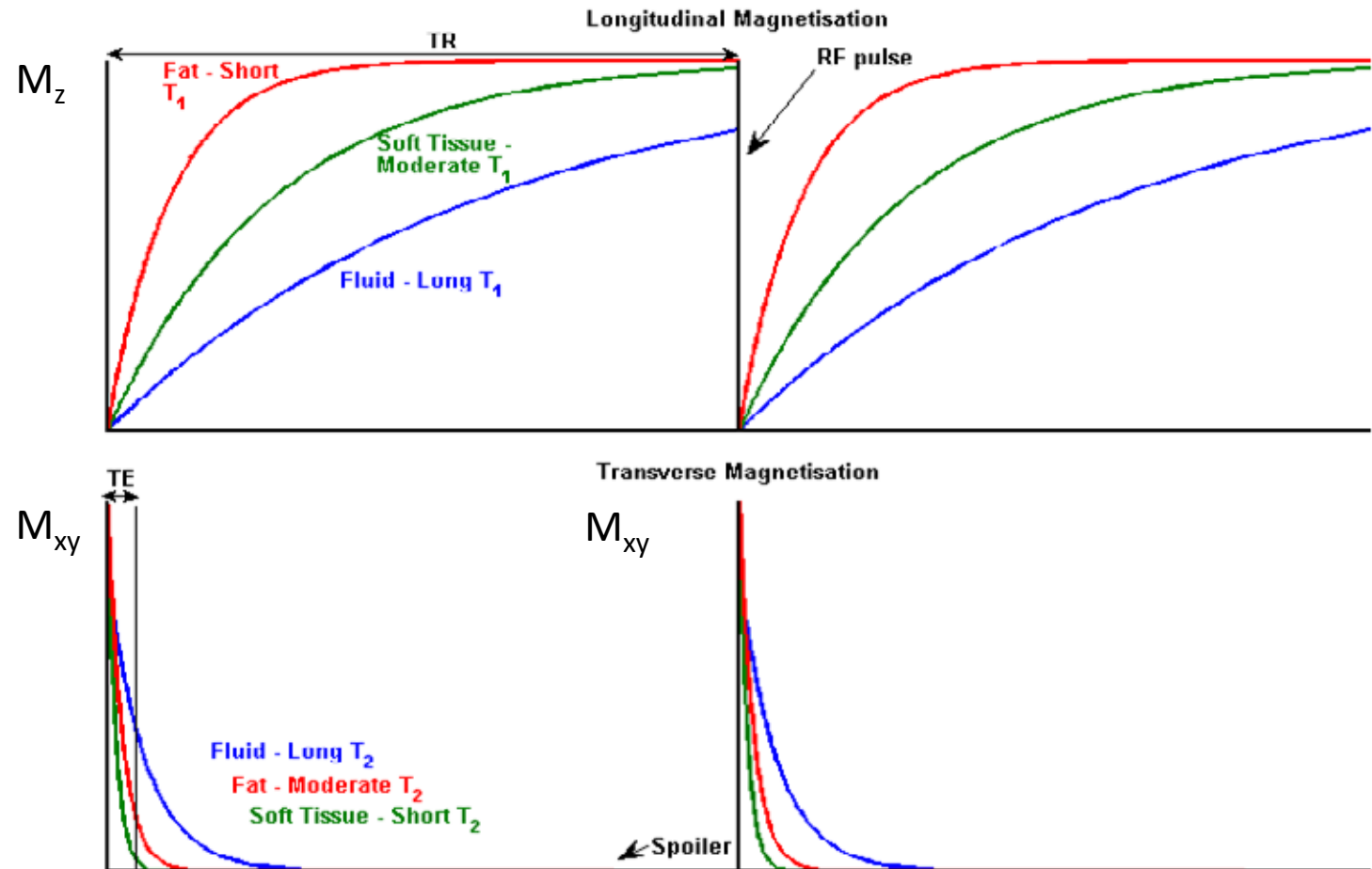
Mechanism

- Proton density weighting is achieved by choosing scan parameters that minimise signal differences due to magnetisation relaxation times.

Magnetisation Evolution

- Long TR minimises differences in longitudinal magnetisation.

- Short TE minimises differences due to transverse decay.

- Signal differences mostly due to differences in proton density.

T_1 weighting

Short TR Short TE

T_2/T_2^* Weighting

Long TR Long TE
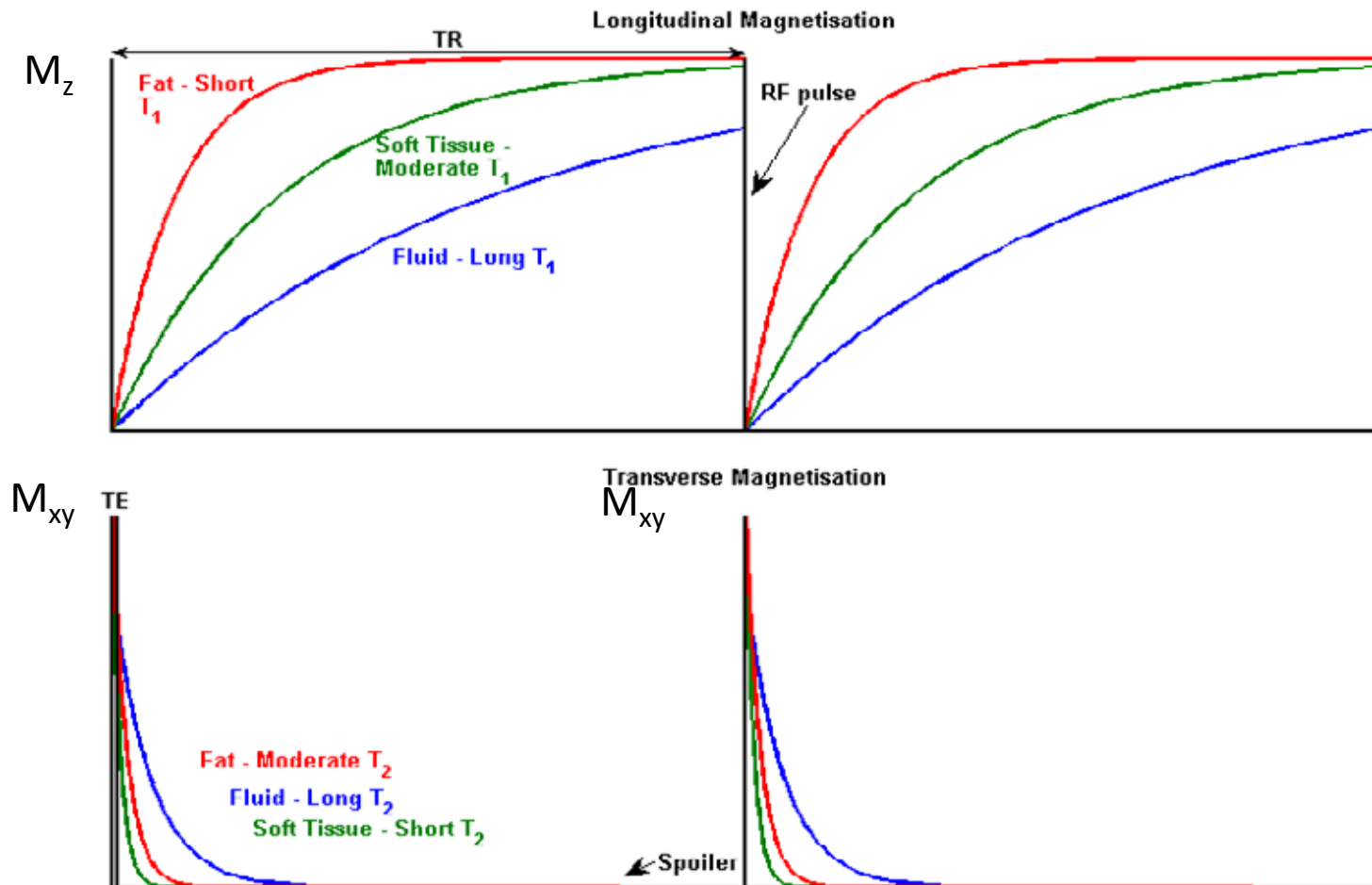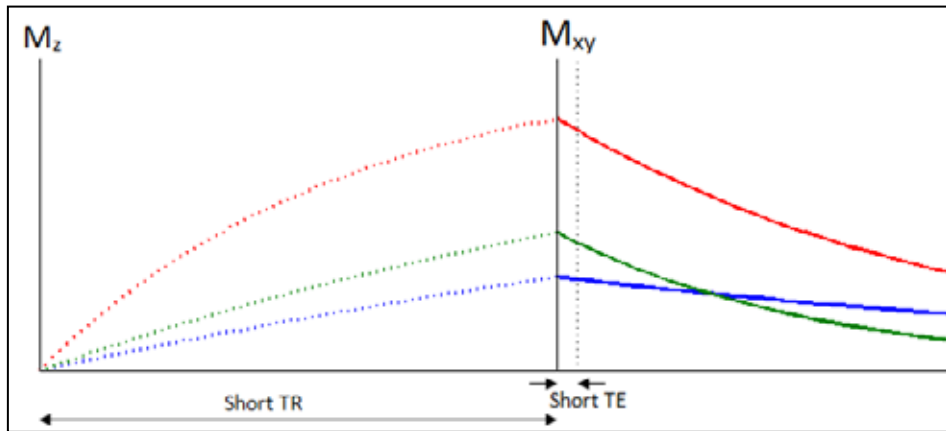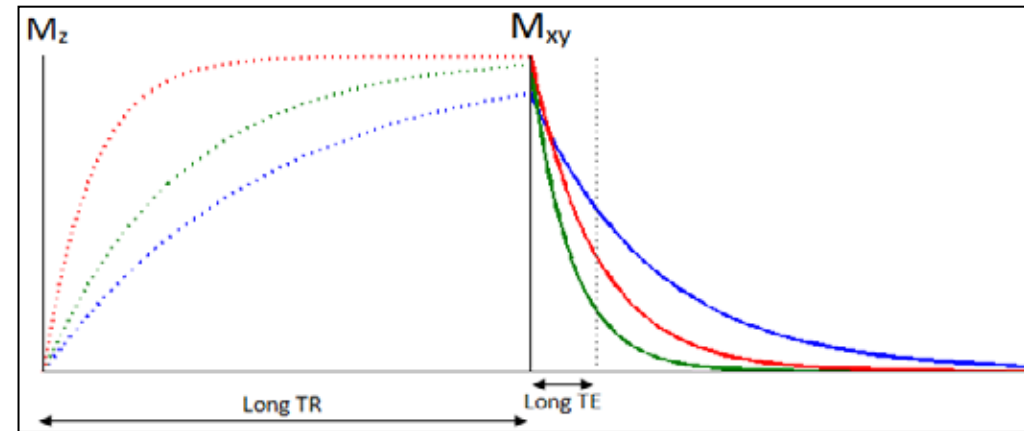
Contrast weighting

- T_1 weighting relies on variable degrees of longitudinal recovery while T_2/T_2^* weighting relies on variable degrees of transverse decay. To achieve proton density weighting both T_1 and T_2/T_2^* weighting are minimised.

Proton Density Weighting
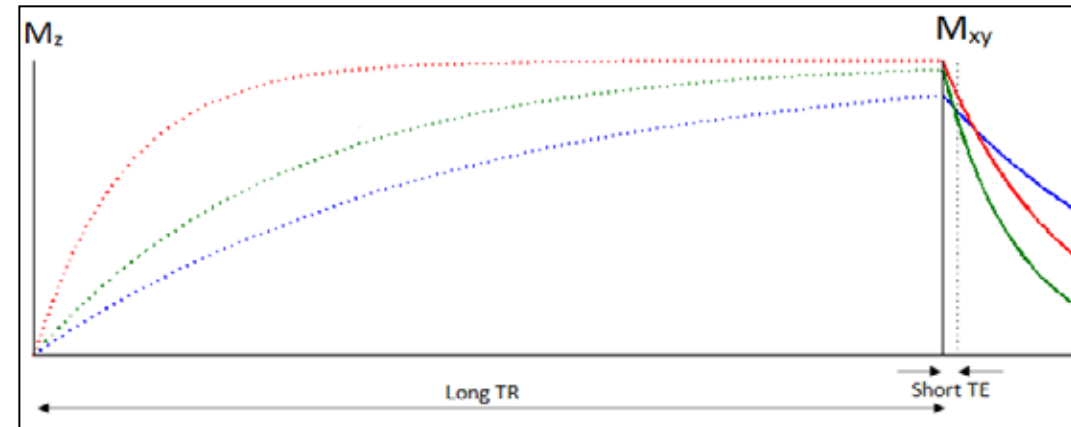
Long TR Short TE

Weighting	Short TR	Long TR
Short TE	T_1	PD
Long TE	-	T_2/T_2^*

Fat – Short T_1 & Moderate T_2/T_2^*
Soft tissue – Moderate T_1 & Short T_2/T_2^*
Fluid – Long T_1 & Long T_2/T_2^*

Motivation

- It is often desirable to shorten TR to reduce scan durations. However, if TR is very short SNR is severely affected as there is not enough time for longitudinal recovery. Gradient echo pulse sequences reduce the readout flip angle so that some magnetisation is maintained in the longitudinal direction, whilst a component of magnetisation is measurable in the transverse plane.

T_1 Weighting And SNR

- A low flip angle reduces T_1 weighting as the overall variation in longitudinal magnetisation is reduced. However, less magnetisation is tipped into the transverse plane so SNR is lower.

- A higher flip angle increases T_1 weighting and, if TR is long enough, increases SNR as a greater proportion of magnetisation is rotated into the transverse plane to generate signal.

- If a high flip angle is used with a short TR the SNR may be too low, as the longitudinal magnetisation does not recover sufficiently between RF pulses.

Longitudinal Magnetisation (M_z)

Low Flip angle

90° Flip angle

Short T1
Medium T1
Long T1
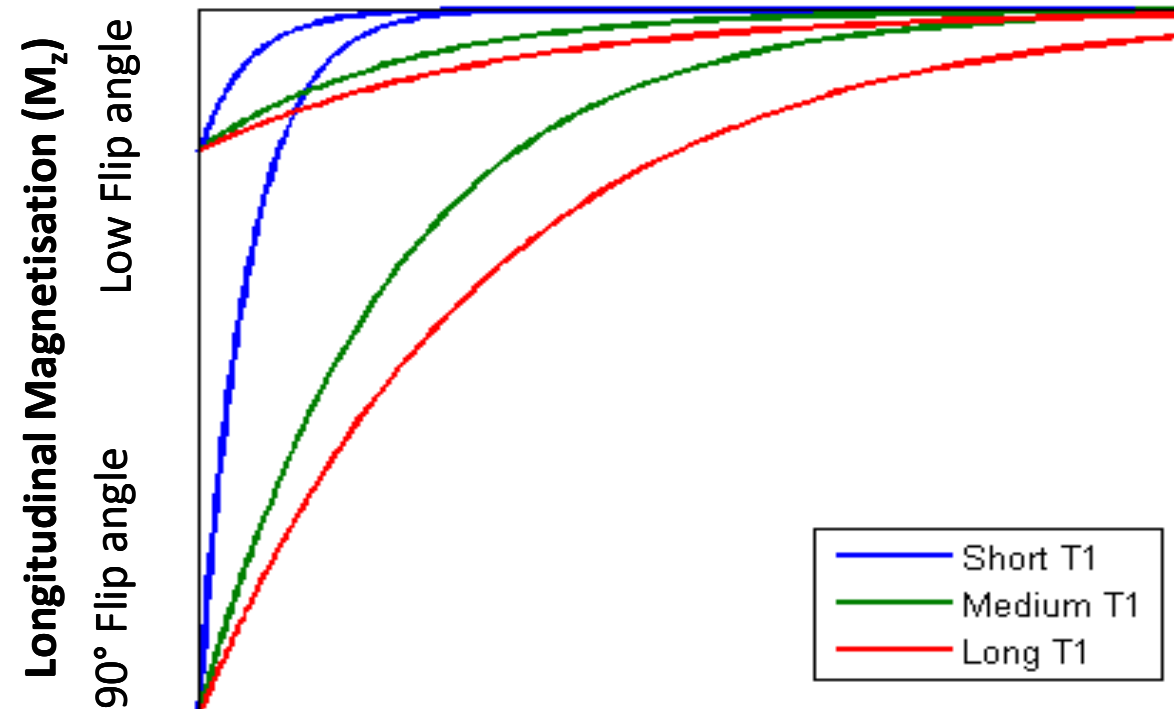
Parameters

- T_2 or T_2^* weighting is achieved by increasing the TE to allow greater relative differences in signal to arise due to different transverse decay times.

- T_1 weighting is achieved by decreasing TR and/or increasing flip angle to increase relative differences in longitudinal magnetisation.

- The signal is also proportional to the relative density of signal generating protons or proton density. A proton density weighted image can be generated by minimising T_1 and T_2 or T_2^* contrast.

- Increasing T_1 and T_2 or T_2^* weighting reduces the signal intensity so increasing weighting leads to a decrease in SNR.

Parameter change	Effect on weighting	Effect on SNR
Increase TE	↑ T_2 or T_2^*	↓
Decrease TR	↑ T_1	↓
Increase flip angle	↑ T_1	TR dependent

Background

- A pulse sequence diagram is a graphical representation that shows the timings of the various components of an MR sequence.

Transmitted Radiofrequency (RF) pulses

- RF pulses with flip angle indicated (α indicates small flip angle (<90°) e.g. for gradient echo).
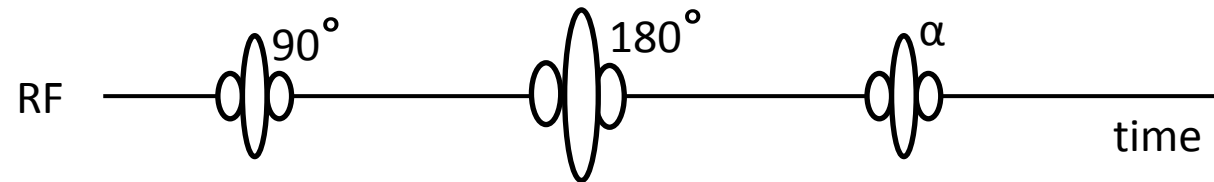
Received Signals

- Echo indicates when analogue-to-digital convertor (ADC) is turned on and signal is received.
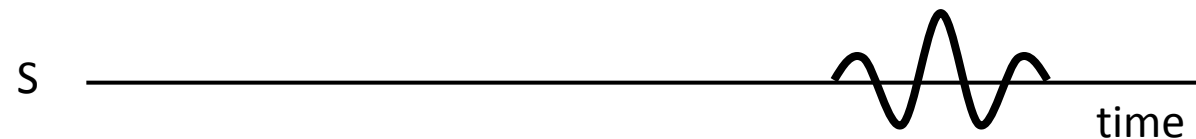
Gradients

- Gradients indicated by trapezoids (sloped edges represents time taken to switch gradient).

- A subscript "dir" indicates direction and may relate to spatial encoding (e.g. slice select (SS), phase encoding (PE) or frequency encoding (FE)) or geometric directions (e.g. x, y or z).

G_{dir}

time

†

*
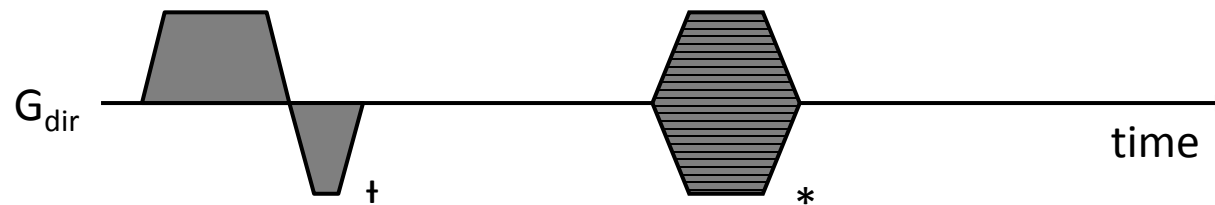
- † Gradients pulses often include a rephasing component with the opposite polarity.

- * Stripes indicate gradient is altered between repetitions (e.g. for phase encoding). As imparting a phase shift is the purpose of the phase encoding gradient pulse these pulses do not include a rephasing component.

Spoiled Gradient Echo

Spoiled Gradient Echo Sequence

- Gradient echo based pulse sequences use a small (<90°) flip angle excitation pulse and bi-polar gradients to generate an echo.

- In order to reduce imaging times a flip angle α < 90° is used. The RF pulse only tips a component of the magnetisation into the M_{xy} direction, which means that the measurable signal will be smaller than for a 90° pulse. However, a significant component of magnetisation remains in the M_z direction which means that the TR required to achieve full recovery of the longitudinal magnetisation can be much shorter. This in turn reduces the total acquisition time.

- To read the signal an initial gradient in the frequency encoding direction is applied (G_{fe}). Spins will precess at different frequencies along this gradient and the transverse magnetisation will dephase. A second gradient is then applied which has the same amplitude as the first but a gradient slope in the opposite direction. In practice this second gradient is applied for twice as long as the first so that the spins rephase into a maximum signal amplitude at the centre of the readout gradient (TE) and then dephase again generating a symmetrical gradient echo.

Timing Parameters

- TR = Repetition Time (excitation to excitation)
- TE = Echo Time (excitation to echo)

Pulse sequence diagram

RF pulse (typically <90°) — RF — α °, α°

Slice select gradient (with re-phasing)
Spoiler gradient prior to next excitation — G_{SS}

Phase encoding gradient — G_{PE}

Readout gradients — G_{FE}

Echo received — S

TR, TE

Spoiled Gradient Echo

Echo Formation

RF

G_{fe}

t

Signal (M_{xy})

time

Signal re-phased by 2nd magnetic field gradient to produce a gradient echo.

T_2 relaxation curve

T_2* relaxation curve

Echo time, TE

Small (α < 90º) Flip Angle

z

α

x

y
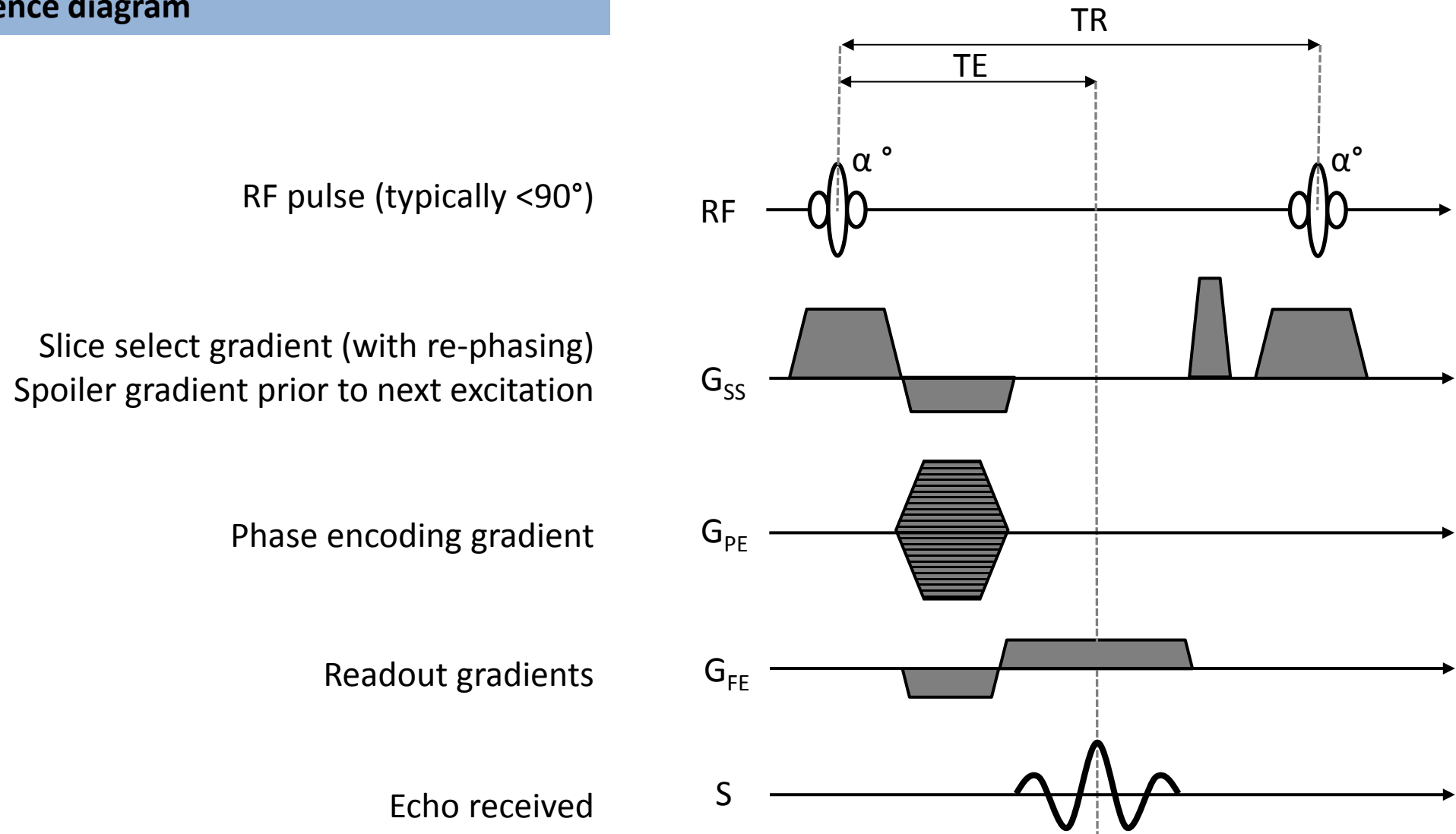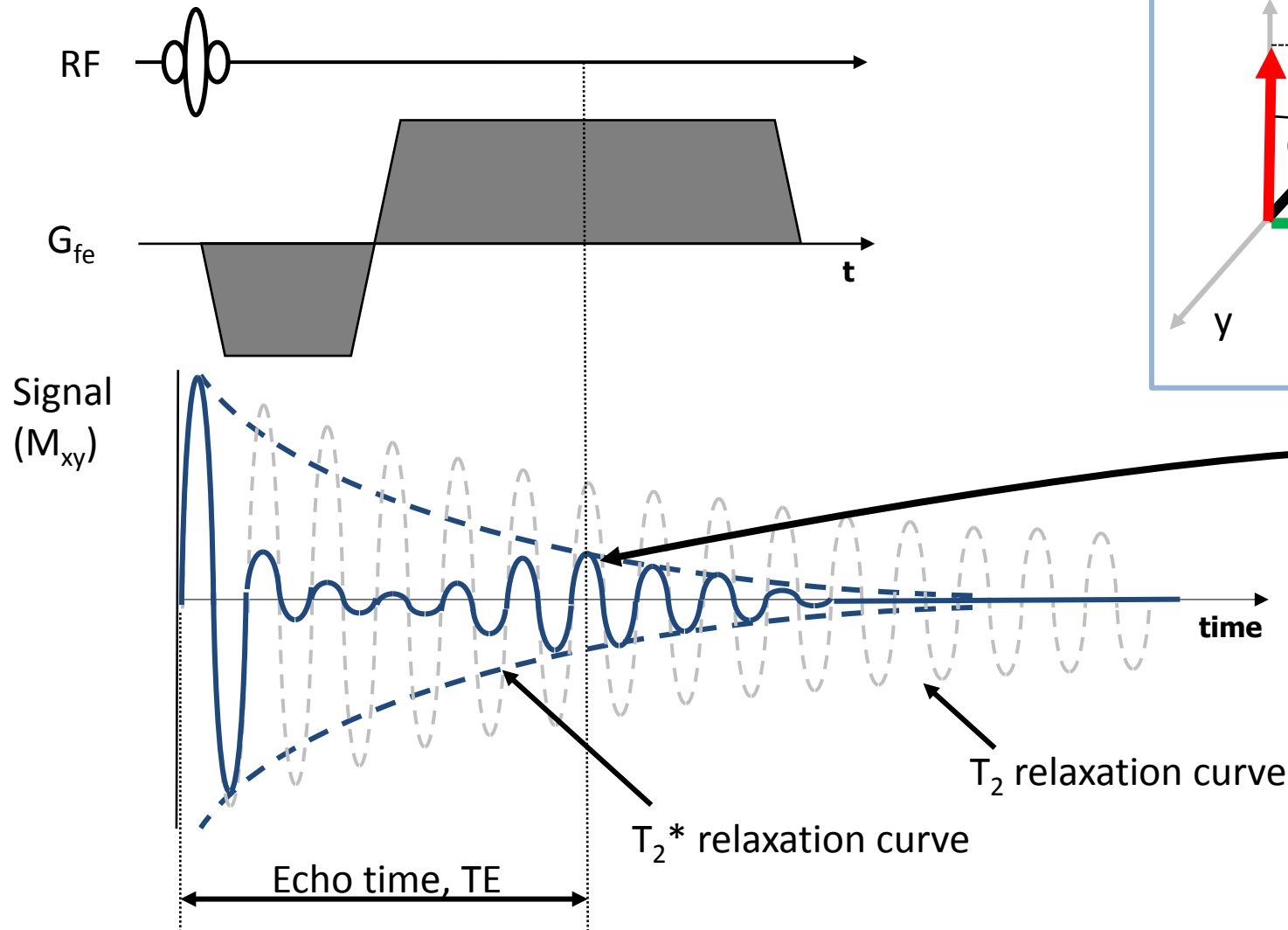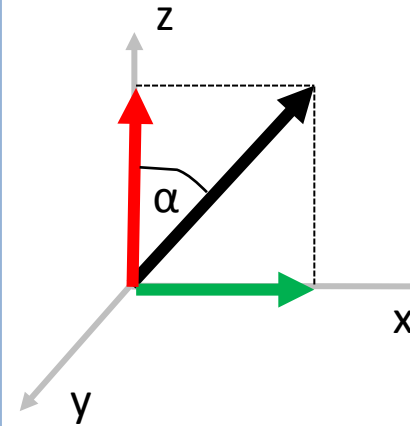
A significant component of M_z remains allowing faster recovery and shorter TR

M_{xy} is reduced, resulting in less signal and poorer SNR compared to a 90° pulse

Spoiler Gradients

- Due to the small flip angles used in gradient echo it is possible to have such a short TR that the transverse magnetisation in the M_{xy} direction has not fully dephased when the next RF pulse is applied.

- In spoiled gradient echo the remnant transverse magnetisation is destroyed before the end of the TR period using a spoiler gradient (also known as a crusher gradient) or a spoiler RF pulse so that it does not contaminate subsequent lines of k-space.

- The dependence of this sequence on bipolar gradient pulses makes it intrinsically sensitive to flow. High velocity gradients in flow jets will cause dephasing due to the range of velocities within a voxel, and will be visible on the images as a signal void.
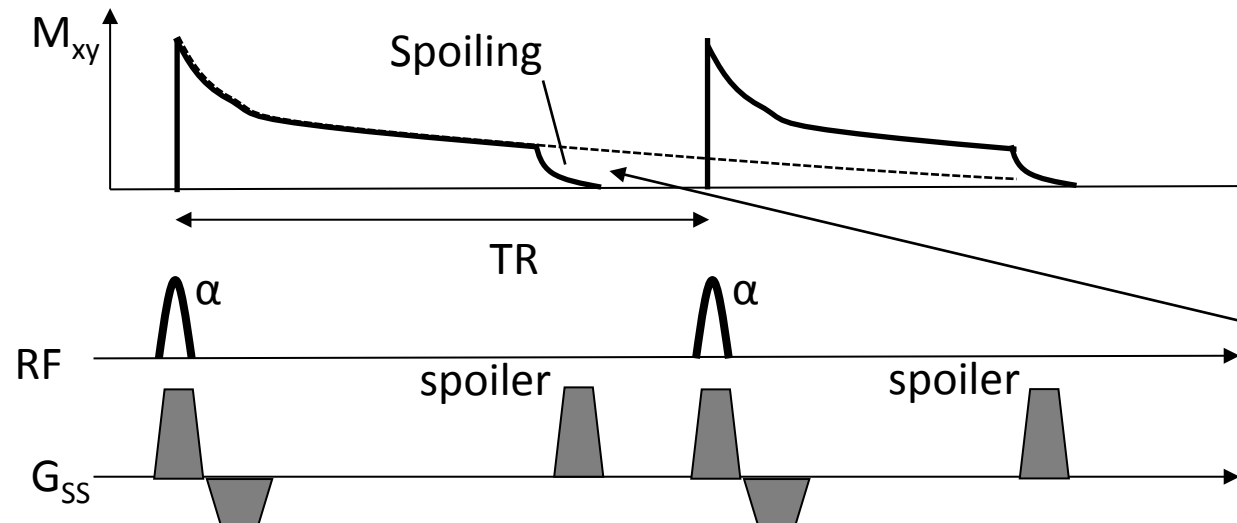
If not spoiled, remnant transverse magnetisation would corrupt subsequent signal acquisitions

Repeated Excitation

- In a spoiled gradient echo acquisition spins are repeatedly excited without allowing time for the longitudinal magnetisation to recover to its equilibrium value. Consequently the longitudinal magnetisation prior to each RF pulse changes after the first excitation.

- If a 90° pulse is used the available magnetisation for the first signal acquired is the maximum possible (M_0) and for subsequent signals it is equal to the longitudinal magnetisation that recovers from zero between readout pulses.

- If a smaller flip angle is used the longitudinal magnetisation prior to each RF pulse reduces throughout the first part of the acquisition, reaching a steady state value after a number of repetitions.
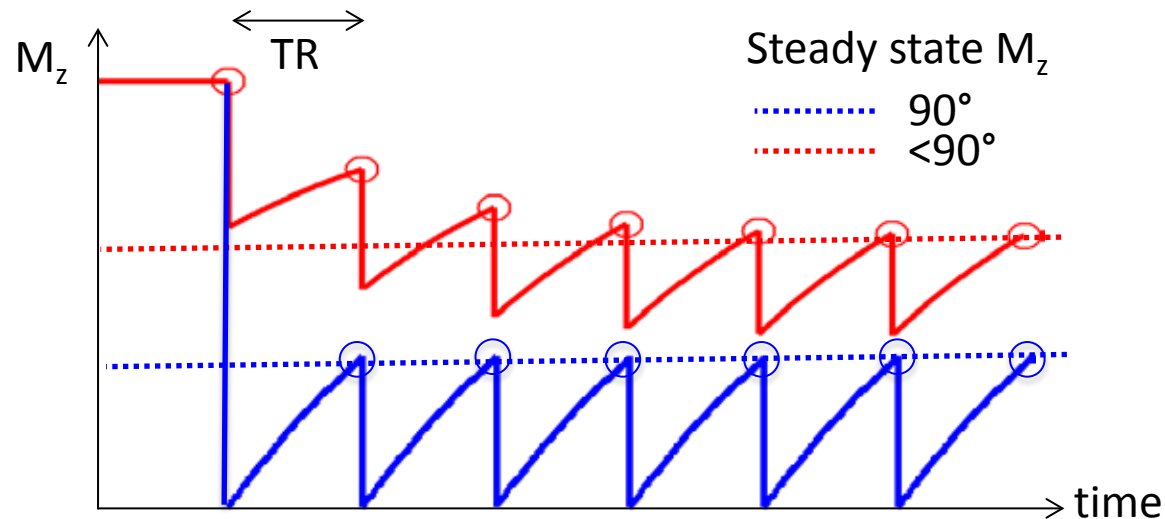
Spoiler Gradients

- When imaging the same tissue repeatedly signal saturation occurs due to incomplete recovery of longitudinal magnetisation between RF pulses.

- After several pulses a steady-state longitudinal magnetisation will be reached.

Fresh Blood

- Longitudinal magnetisation in stationary tissue will be reduced due to saturation.

- However, blood flowing into the imaging volume has not experienced the prior train of RF pulses so is not saturated. Consequently the longitudinal magnetisation of blood flowing into the volume is larger than that in stationary or slow moving tissue.

- Blood therefore appears bright in GE images due to this inflow related enhancement.

Excitation RF Pulse	Delay (duration TR)	Next Excitation
Spins in stationary tissue are partially saturated due to repeated excitation	Blood is continuously replaced so does not become saturated	Steady-state longitudinal magnetisation is greater in blood giving higher signal
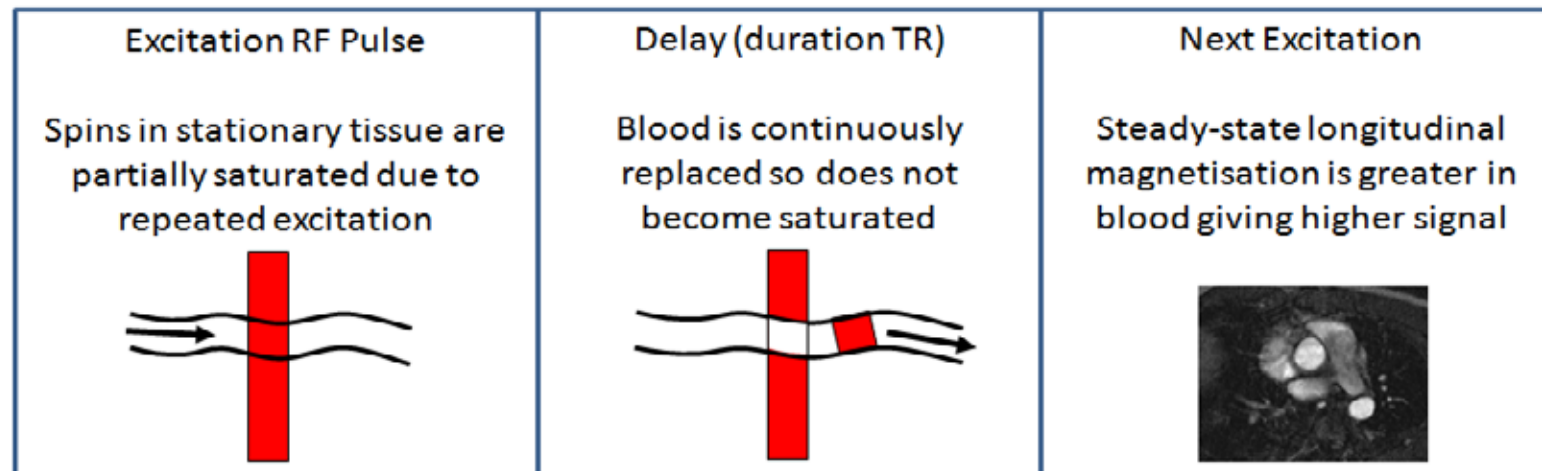
Non-Gated Techniques

- Significant respiratory motion whilst imaging will cause ghosting artefacts.

- The ideal solution is to image so quickly that significant movement does not occur during imaging. However, such short acquisition times severely limit the potential resolution and SNR.

- A common solution to respiratory motion is to acquire images whilst the patient holds their breath. If the patient complies well with breath-holding instructions this provides a 10-15s window where images can be obtained without respiratory motion. However, some patients cannot comply with breath-holding instructions and some imaging scenarios require acquisition times that are longer than a typical breath-hold.

Respiratory Gating

- Respiratory gated techniques monitor respiratory motion and only acquire data within a predefined window.

- An alternative is to use MR navigator echoes. A column of tissue is excited before acquiring each image using a specially designed RF pulse. The column is positioned perpendicular to the diaphragm so that when the signal is reconstructed to form a one dimensional projection image a clear bright to dark boundary is visible between the lungs and the liver. The position of the boundary can then be detected and used as a measure of the diaphragm position.
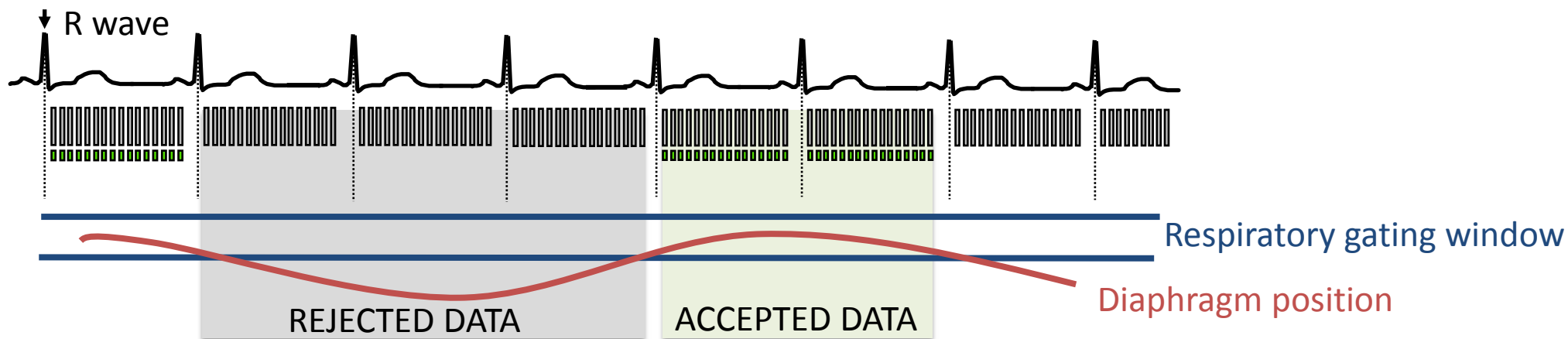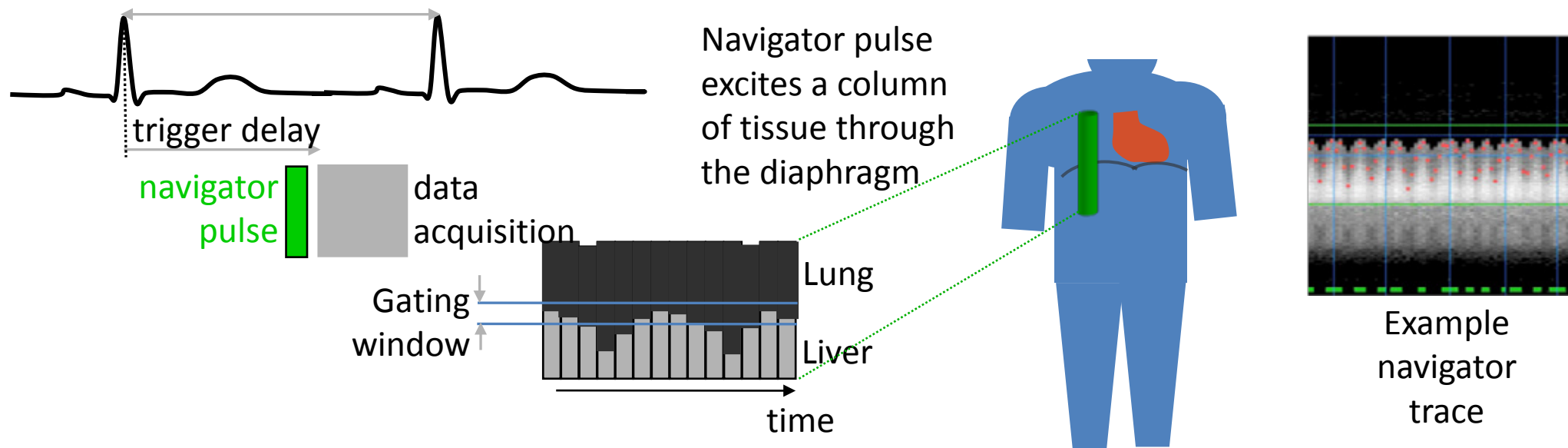
Cardiac Motion

- During the cardiac cycle the heart performs a complex motion pattern including contraction, twisting and shortening.

- In addition to the motion of the myocardium there is also variable, high velocity flow of the blood in the heart in multiple directions.

- This combination of soft tissue movement and fluid flow makes imaging in the heart challenging. Without accommodating for this motion images of the heart would contain several artefacts and have limited clinical utility.

- Consequently cardiac imaging is synchronised to the heart beat and images are acquired as quickly as possible to freeze motion. This is normally achieved through use of an electrocardiography (ECG) system which integrates with the scanner interface.

ECG Synchronisation

- MR compatible leads are attached to the patient before imaging.

- Software analyses the ECG trace to find the QRS complex and generates a synchronisation pulse.

- This initiates the pulse sequence controller so that the pulse sequence is applied at a given time after the R wave, known as the trigger delay.
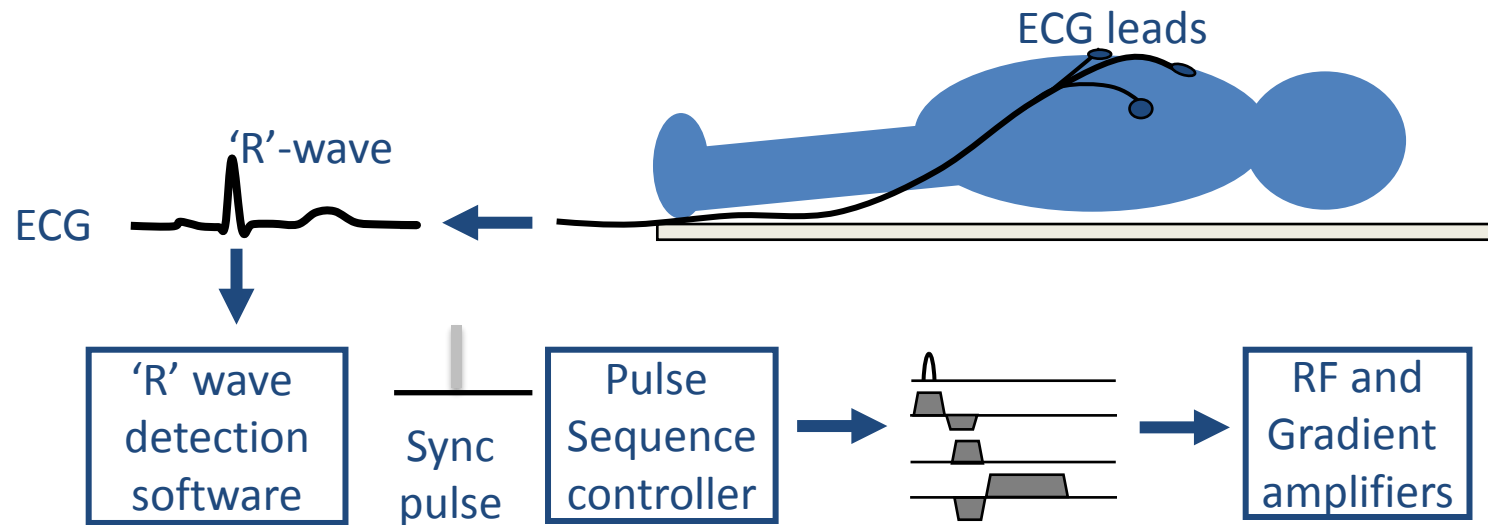
Conventional ECG Triggered Pulse Sequence

- In a conventional ECG triggered image acquisition one line of k-space is acquired per R-R interval.

- The trigger delay sets the phase of the cardiac cycle that is to be imaged (short trigger delay – systole, long trigger delay – diastole).

- The TR must be a multiple of the R-R interval because the time between acquisitions is set by the heart rate.

- The technique generates a single image at a single point in the cardiac cycle. High resolution imaging is possible because the image is constructed over multiple cardiac cycles. However, scan times will be long (TR x number of k-space lines).

- In practice conventional imaging of a single cardiac phase as described here is not performed because of the large proportion of time wasted whilst no data is being acquired.

Conventional ECG Triggered Pulse Sequence

ECG

1

2

3

4

Trigger delay

TR = R-R interval

1
.
.
.
.
.
.
n

Data Acquisition

1 2 3 4 n

Scan time

Scan time = n x TR
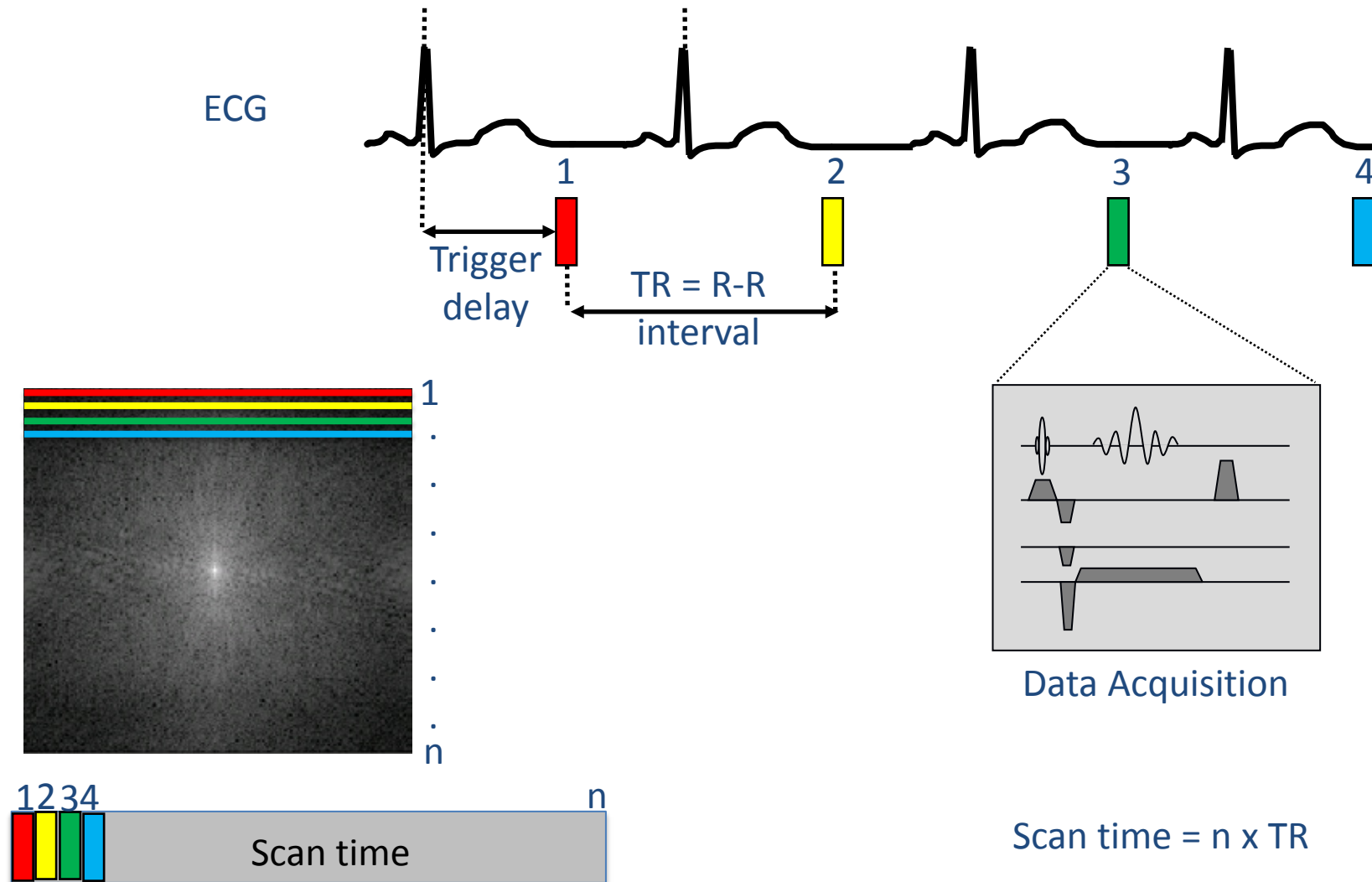
Method

- Cine imaging produces a movie showing a single slice of the heart for each phase of the cardiac cycle.

- Data acquired at each phase of the cardiac cycle is assigned to a separate k-space corresponding to that cardiac phase.

- The full k-space acquisition is built up over a number of R-R intervals.

- The resultant images reconstructed from each k-space are then displayed in a movie loop.

- The technique requires short TRs and therefore can only be achieved with gradient echo sequences.
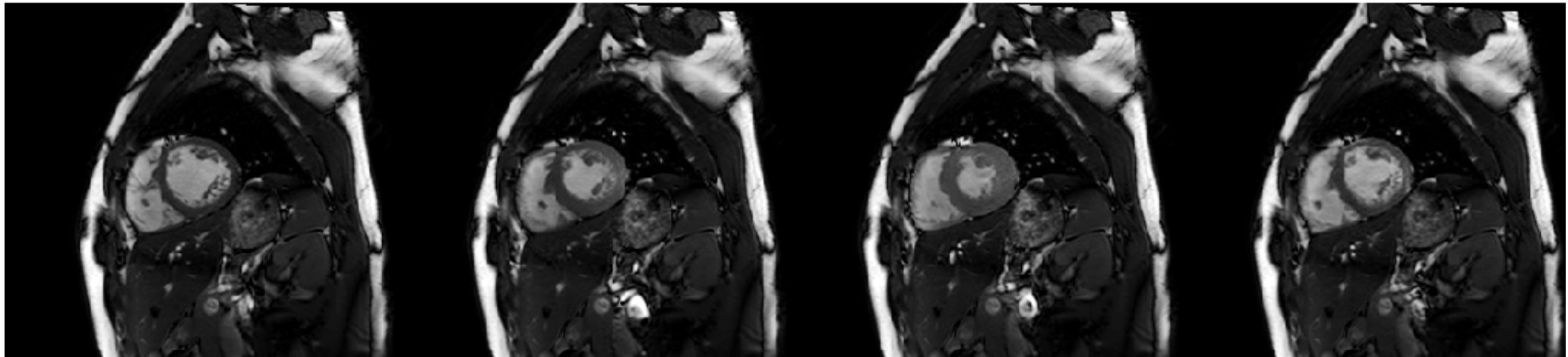
cardiac phase

Cine Imaging

Cine Imaging With Prospective Triggering

ECG triggering

Cardiac phase number

1 2 3 4   1 2 3 4   1 2 3 4   1 2 3 4

TR

dead space

Retrospective Gating

ECG gating

TR

Cardiac phase number

1 2 3 4 5   No dead space
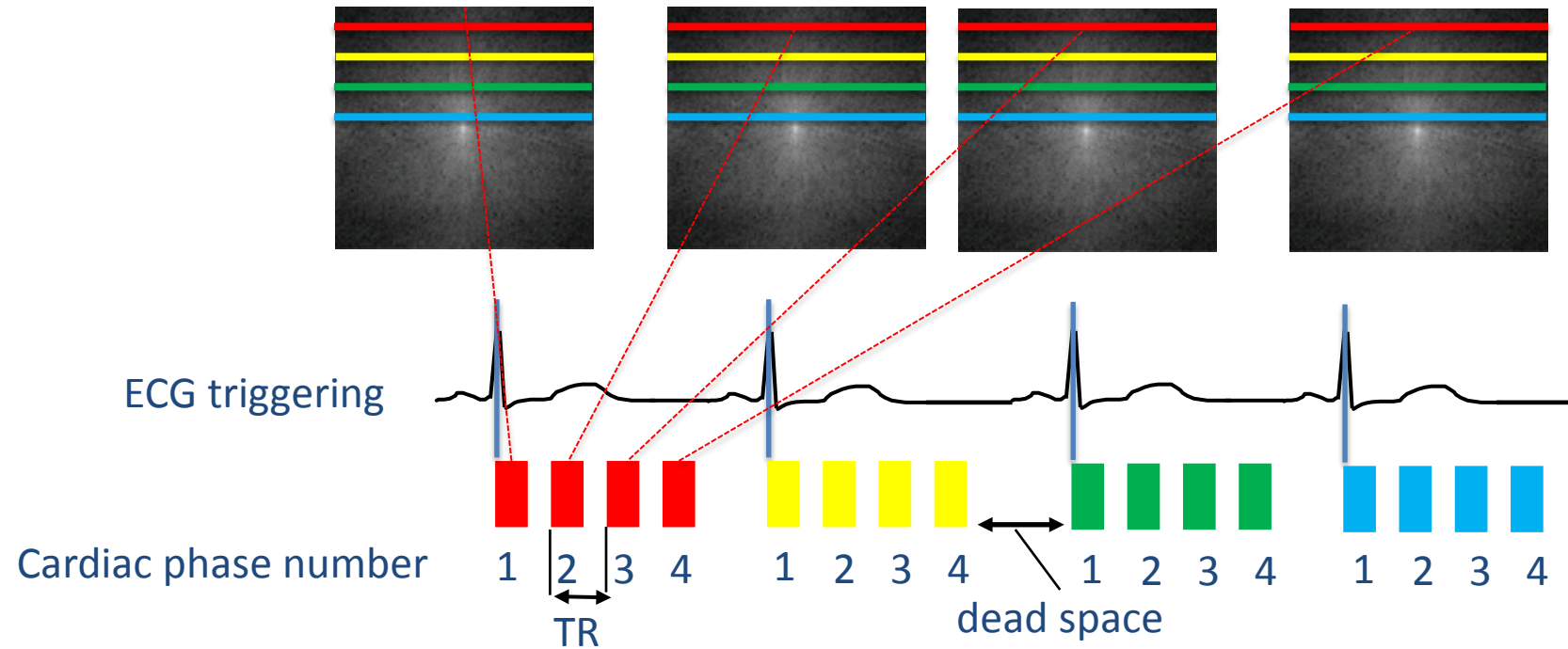
Cine Imaging
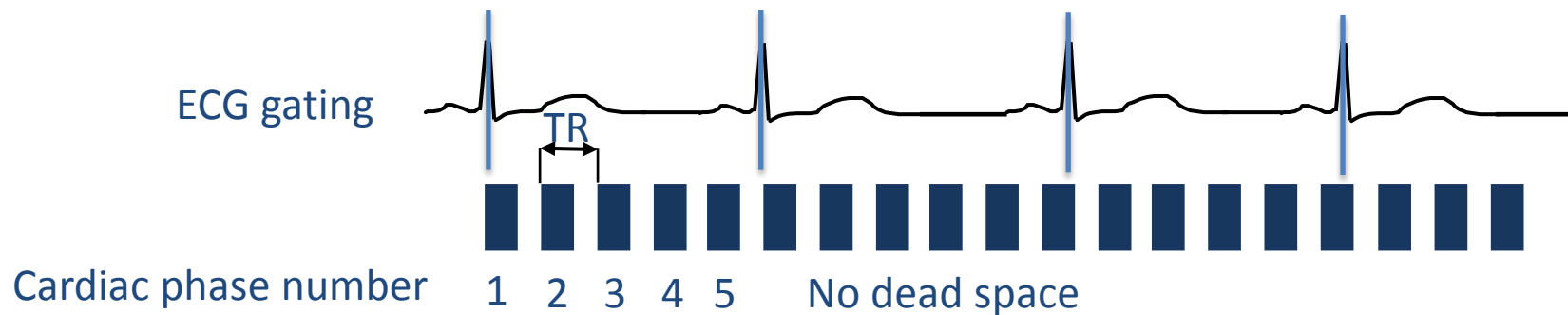
Prospective Triggering

- In prospective triggering an estimate of the average R-R interval is made prior to imaging to determine the number of cardiac phases to image.

- Data collection for the first phase is triggered by the R-wave identification and is followed by collection of data for the subsequent phases.

- After the last phase is acquired data acquisition is stopped before the next R-wave trigger resulting in a 'dead space' where data is not acquired.

Retrospective Gating

- An alternative to prospective triggering is retrospective gating, in which data is acquired constantly throughout the cardiac cycle.

- During image reconstruction, a retrospective average heart rate is calculated and the data points from longer and shorter RR-intervals are interpolated onto the average RR-interval so that the image data is mapped onto a pre-determined number of cardiac phases.

- All cardiac phases are therefore imaged, removing the 'dead space' problem.

- Retrospective gating is problematic if there are large beat to beat variations in the R-R interval. In cases where there a large number of arrhythmias retrospective gating is not practical and triggering should be used.

Analysis

- Visual analysis of cine images can identify wall motion abnormalities throughout the cardiac cycle.

- Signal voids caused by high velocity gradients or turbulent flow can also be viewed in the blood pool to qualitatively assess blood flow patterns through the heart. This can help identify stenoses, regurgitation or flow jets.

- By contouring endocardial and epicardial surfaces throughout the cardiac cycle quantitative functional parameters, such as ventricular volumes and ejection fractions, can be calculated.
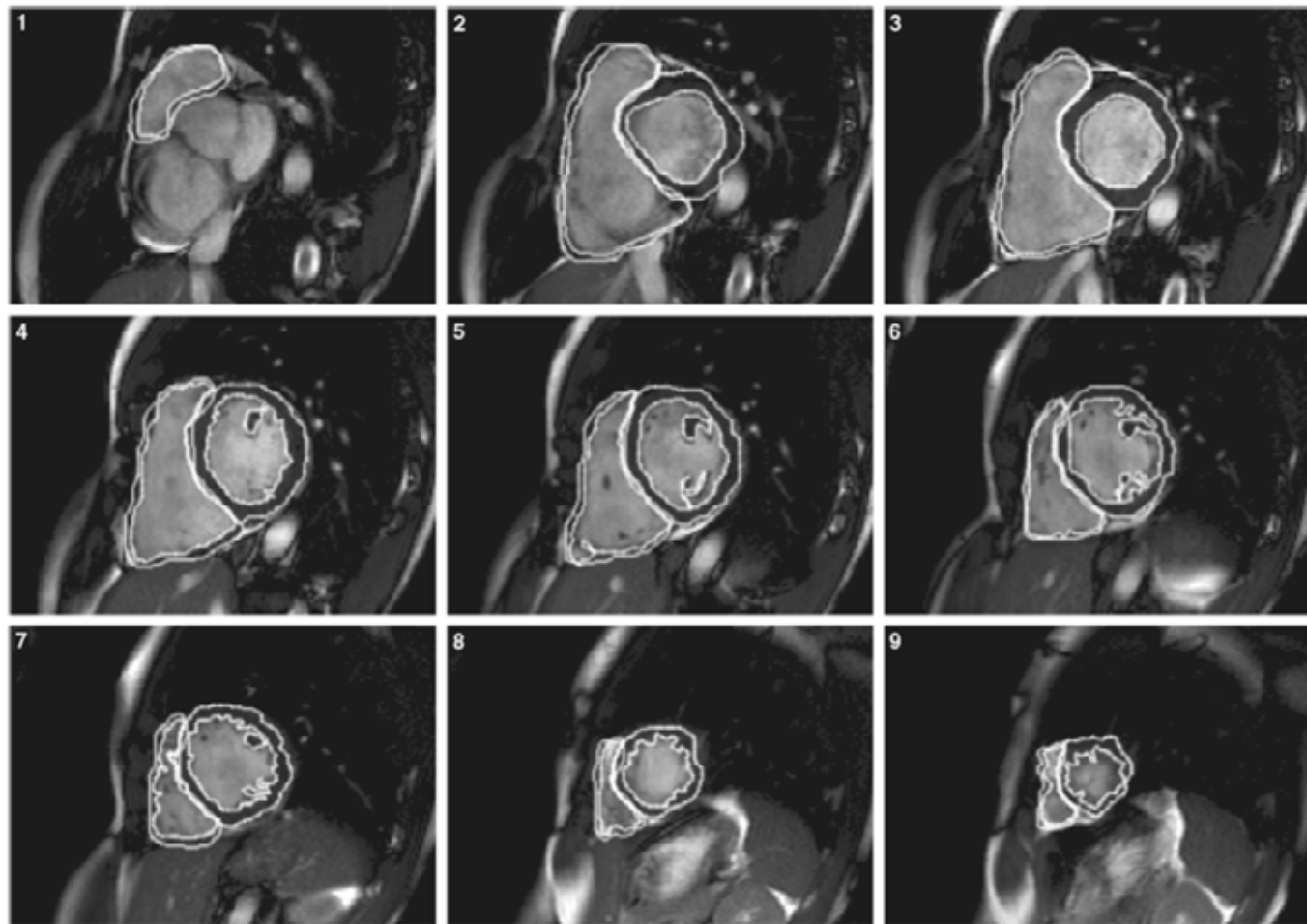
Fast/Turbo Gradient Echo – Segmented Acquisition

- Whereas conventional gradient echo acquires a single line of k-space each R-R interval, fast (also known as turbo) gradient echo acquires multiple lines.
- The low flip angle RF pulse is rapidly repeated to generate a number of gradient echoes with different phase encoding gradient strength so that a number of lines of k-space are filled every R-R interval.

- The turbo factor (TF) describes how many lines of k-space are acquired in a single 'shot' within one R-R interval.

- For a k-space of N_{PE} lines the scan time for conventional imaging would be N_{PE} x TR. TR is equal to the R-R interval.

- Fast/turbo gradient echo reduces the scan time by a factor of TF, so that the new scan time is (N_{PE} x TR)/TF.

Fast/Turbo Gradient Echo

Conventional Gradient Echo

One line per
R-R interval

Scan time = N_{PE} x TR

Fast/Turbo Gradient Echo (Segmented)

Turbo Factor (e.g. TF=4)
lines per R-R interval

Scan time = (N_{PE}/TF) x TR
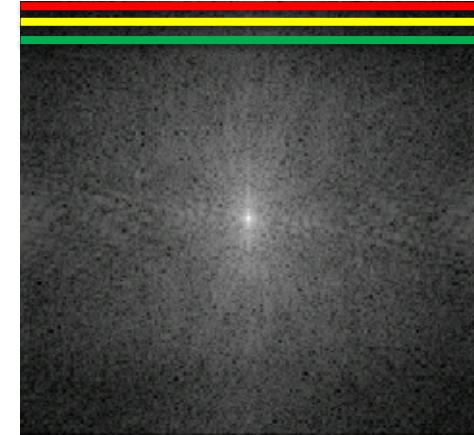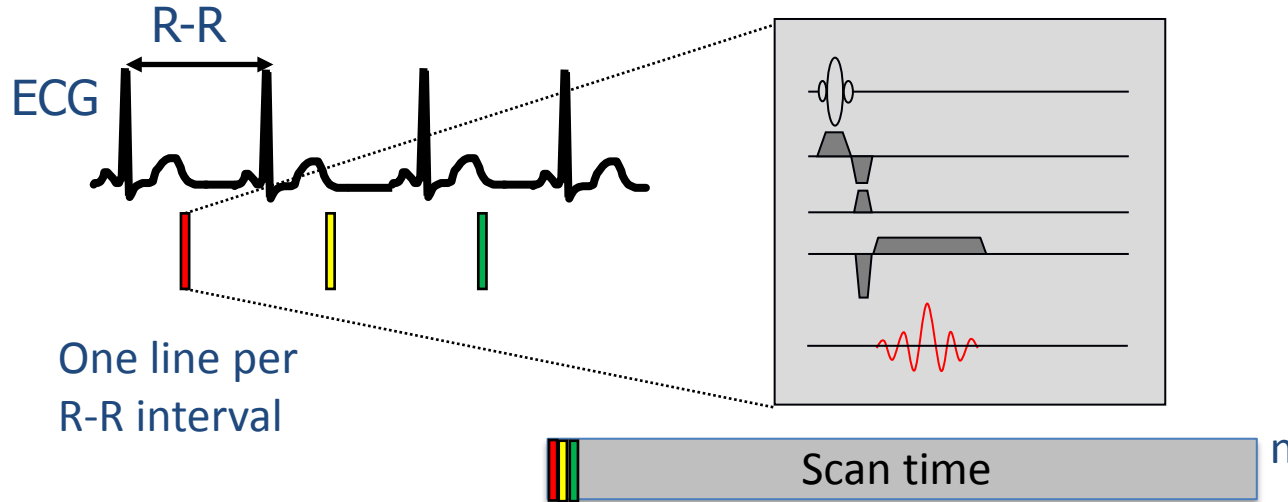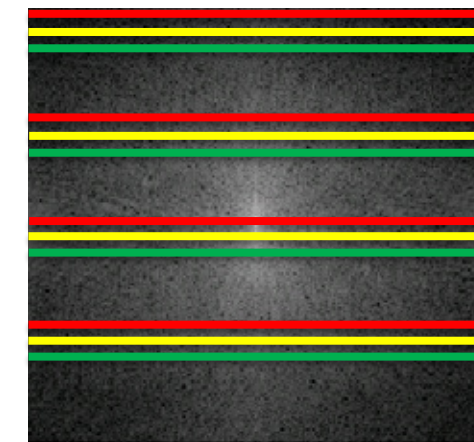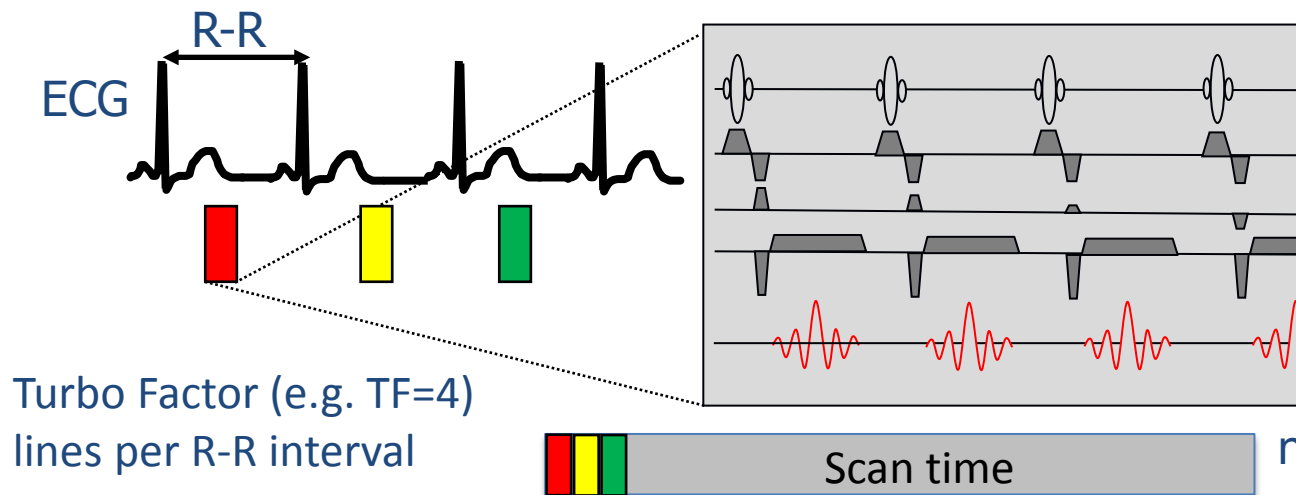
bSSFP

- Due to the low flip angles used in gradient echo it is possible to have such a short TR that the transverse magnetisation in the M_{xy} direction has not fully dephased when the next RF pulse is applied. In balanced SSFP this remnant magnetisation is not spoiled but maintained using additional rephasing and dephasing gradients.

- Dephasing due to every positive gradient is 'undone' by an equal negative rephasing gradient. This includes the addition of a rewinder gradient that reverses the effect of the phase encoding gradient.

- The signal is acquired at the exact midpoint of the sequence so that TE=TR/2, which means that the image contrast is related to the T_2/T_1 ratio.

- The coherent signal carries over to subsequent repetitions and is superimposed onto the transverse magnetisation generated by subsequent RF pulses.

- After a number of repetitions this gives rise to a steady state where the signals from multiple repetitions combine to give a much larger signal (and therefore better SNR than spoiled gradient echo).

Features Of bSSFP

- bSSFP requires accurate, patient specific shimming. If the magnetic field is not uniform the rephased transverse magnetisation from previous TRs can destructively interfere with the newly generated signal creating dark banding artefacts in the image.

- The balanced gradients used by bSSFP significantly reduce dephasing due to flow, so bSSFP sequences are less flow sensitive than spoiled gradient echo sequences.

bSSFP Pulse Sequence Diagram

TR

TE=TR/2

G_{SS}

Dephasing gradient

G_{PE}

Rewinder gradient

G_{FE}

S
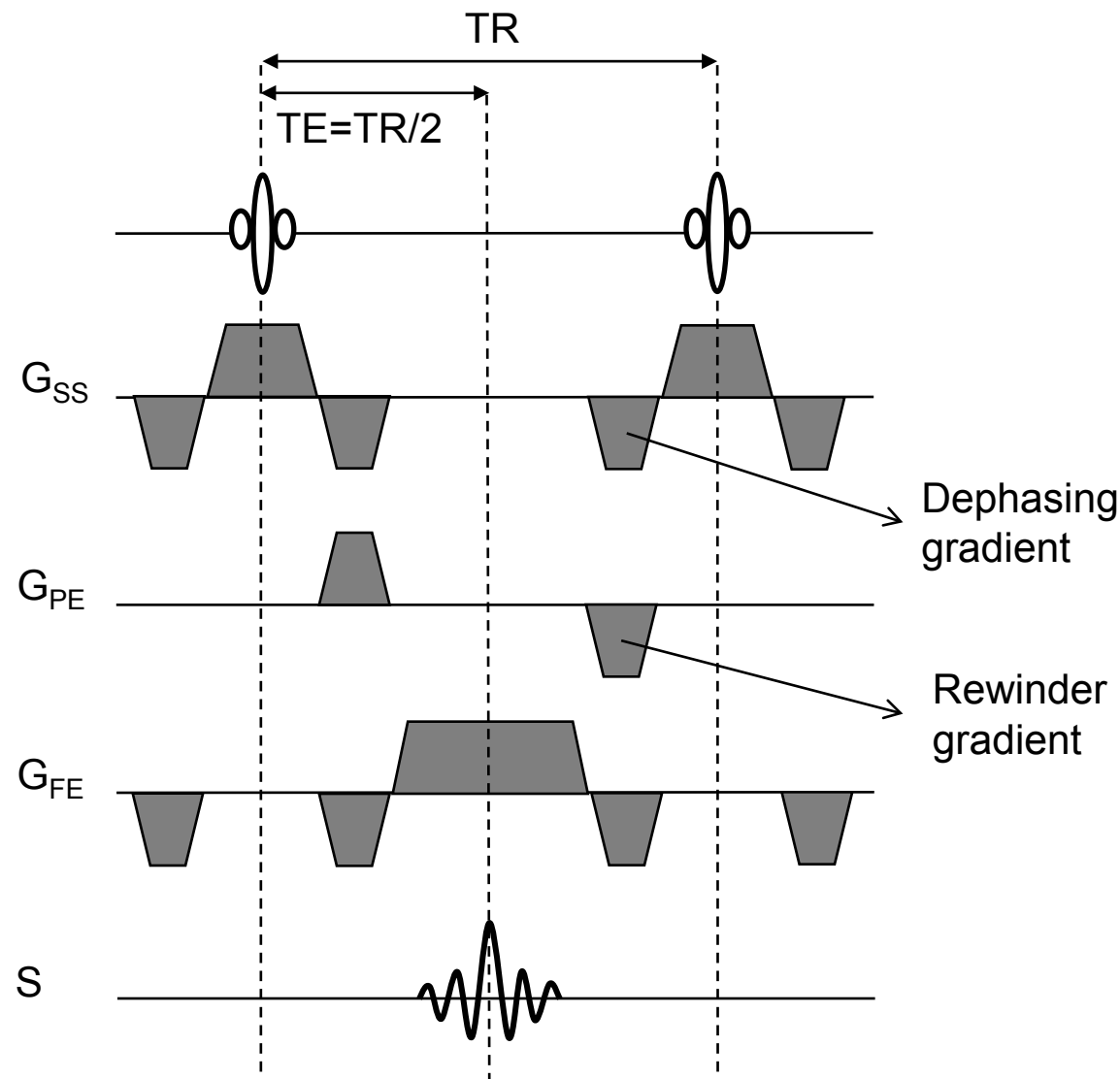
Spoiled and Balanced Sequences

- Two different types of gradient echo sequence are used in cine imaging: spoiled gradient echo and balanced steady state free precession (bSSFP).

- In spoiled gradient echo the remnant transverse magnetisation is destroyed before the end of the TR period using a spoiler gradient or RF pulse so that it does not contaminate subsequent lines of k-space.

- In bSSFP the remnant magnetisation is preserved using a balanced gradient scheme so that remnant magnetisation is superimposed on subsequent read-outs, improving SNR.

- bSSFP relies on a homogenous magnetic field and so good dynamic shimming (using gradient coils to achieve a uniform magnetic field) is essential. Adequate shimming can be harder to achieve at higher field strengths which can limit the use of bSSFP.

Image property	Spoiled Gradient Echo	bSSFP
Blood to myocardium contrast	Variable – relies on through plane flow	Good – T_2/T_1 ratio
Flow sensitivity	High – can visualise flow jets	Low – uses balanced gradients
Image quality	Low SNR – no shimming required	High SNR – good shimming required

Single Shot Echo Planar Imaging (EPI)

- EPI generates multiple gradient echoes from a single RF excitation pulse.

- After an initial RF pulse alternating amplitude gradients are used to repeatedly rephase the signal.

- All lines of k-space are read rapidly in a single shot.

- 'Blipped' phase encoding gradients are used in-between alternating sign frequency encoding gradients to 'scan' k-space as quickly as possible.

- EPI imaging can allow substantially faster image acquisition. However the technique leads to susceptibility weighting and artefacts can arise if errors in phase build up.
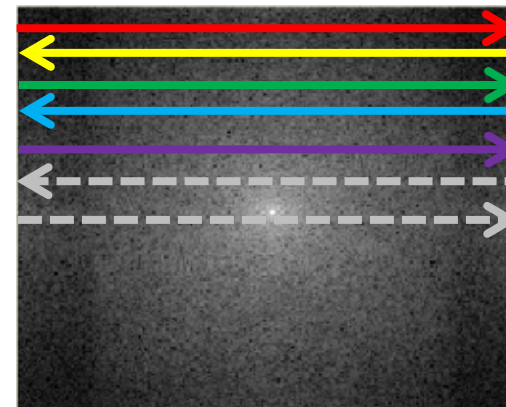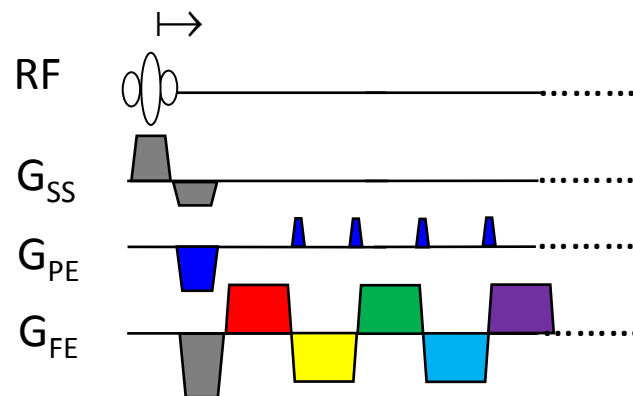
Hybrid (Segmented) EPI

- In practice T_2* dephasing means that there is insufficient signal to read all of k-space from a single RF pulse.

- Hybrid EPI is a half-way approach between EPI and spoiled gradient echo. An echo train length (ETL) of lines of k-space are read using EPI before a new RF pulse is applied to generate the next signal.
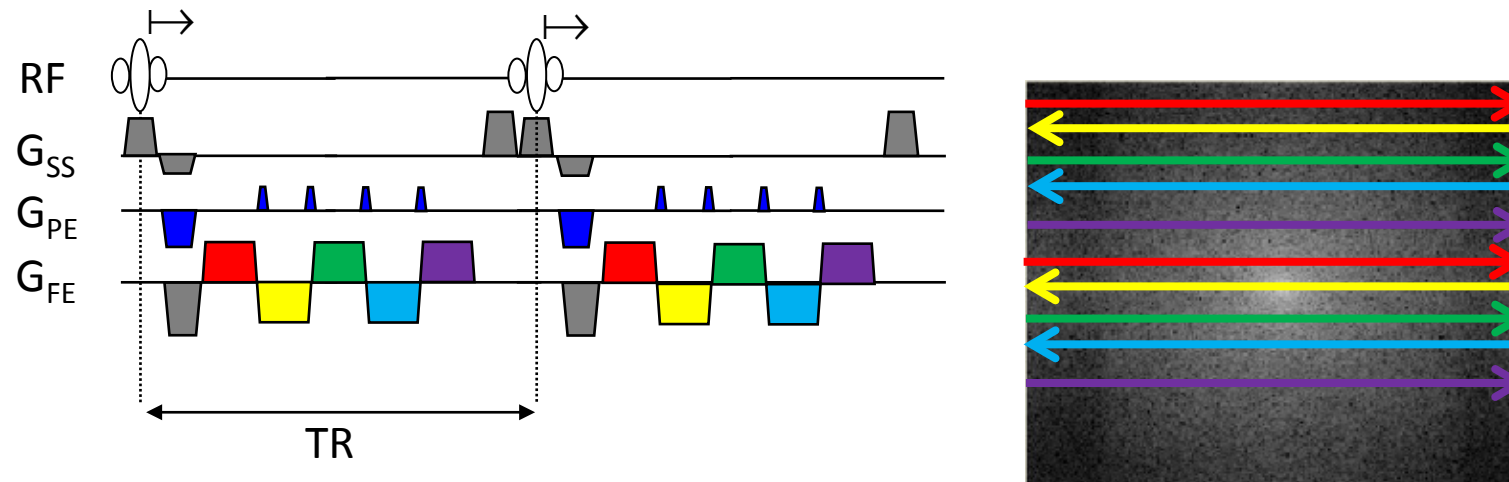
Echo Train Length (ETL) = 5

Effective Echo Time In EPI

- The TE, and so the T_2 contrast, for each read-out will be slightly different.

- The T_2* contrast for the image is dominated by the TE of the central line of k-space, which corresponds to the lowest spatial frequency. This is described as the effective TE (TE_{eff}).

Central k-space line (k_0) dominates image contrast
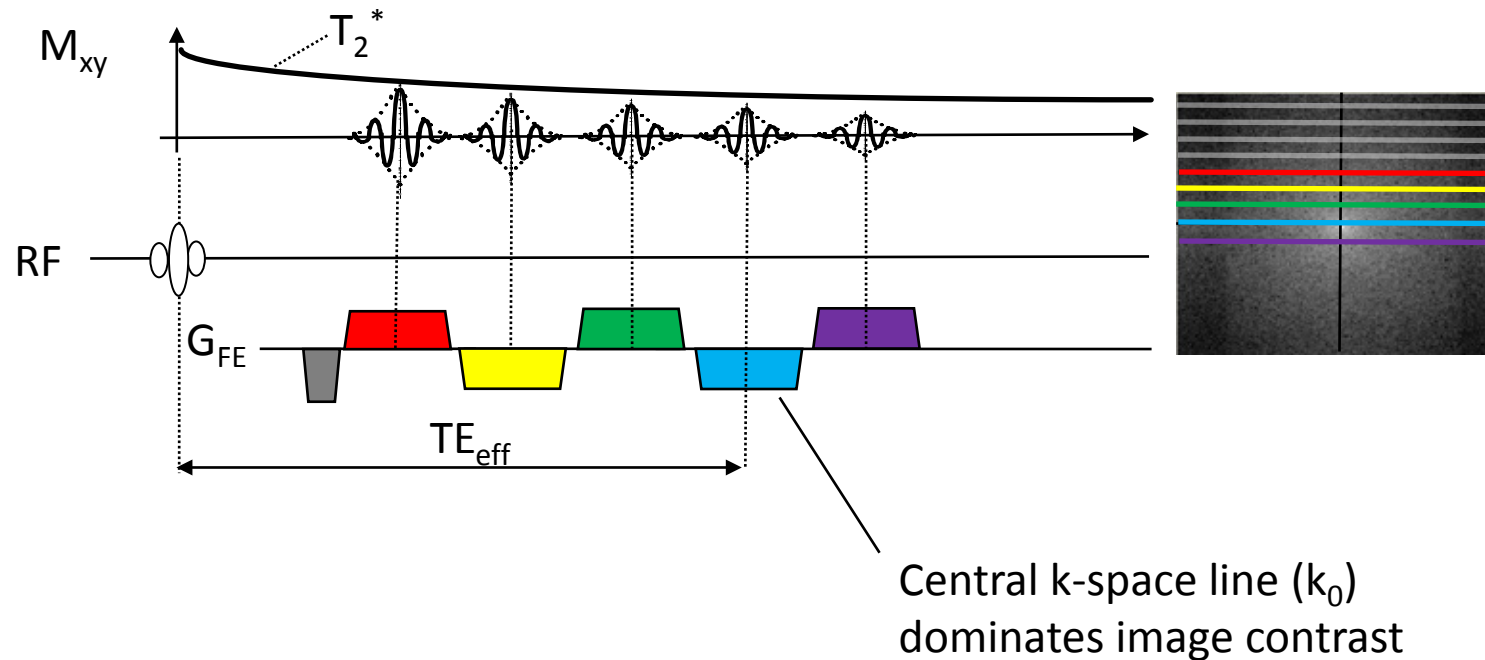
Preparation Pulses

- Due to the small flip angles utilised in gradient echo imaging the signal contrast between tissues with different T$_1$ values is small and image contrast is poor.

- If larger flip angles are used a higher contrast can be achieved but an increase in TR is required to allow for sufficient recovery of the longitudinal magnetisation, increasing scan times.

- To maximise contrast whilst maintaining short scan times an initial preparation pulse (typically a saturation pulse) is used, followed by a preparation pulse delay (PPD, also known as saturation time, TS). This establishes a large difference in net magnetisation which is stored along the longitudinal axis.

- After the initial long preparation pulse delay a standard fast gradient echo read-out train is employed to read the T$_1$-prepared signal out quickly.

- The preparation pulse is typically either a 90° saturation pulse or a 180° inversion pulse although other flip angles may be used.

Fast GE - No Preparation Pulse
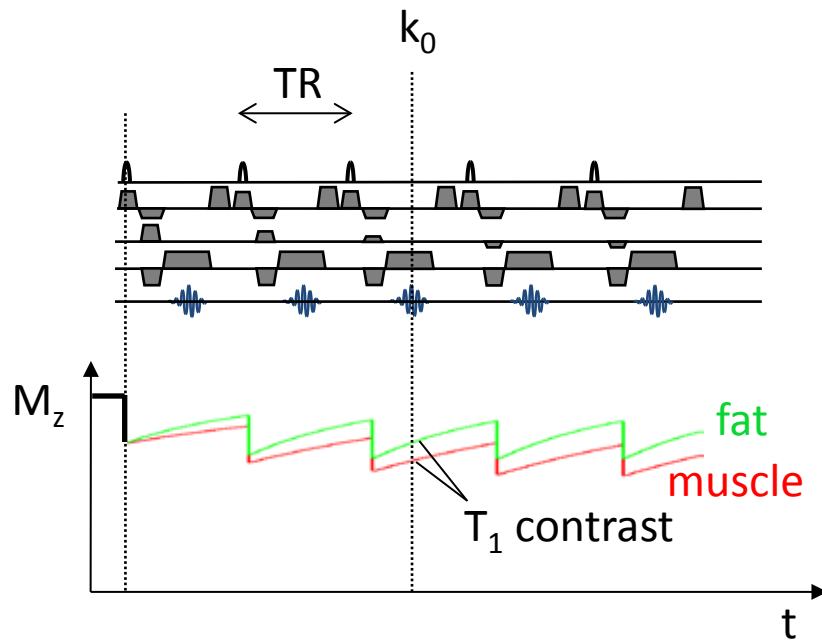
Fast GE – Saturation Prepared
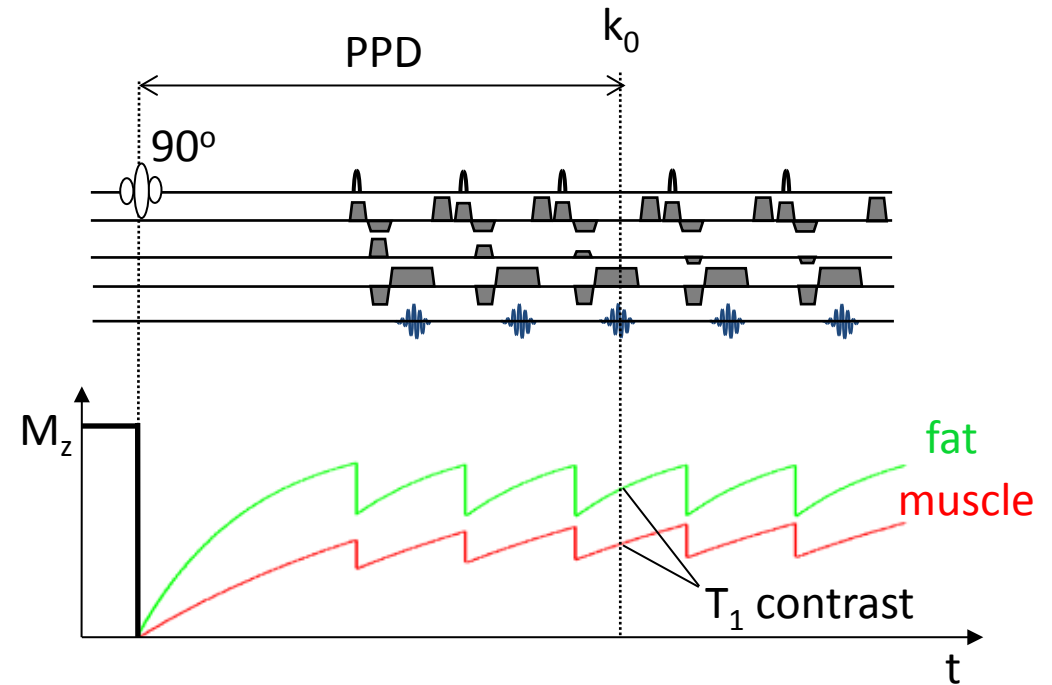
Spin Echo Sequence

- Spin echo uses a 180° refocusing pulse to reverse the effects of inhomogeneities in the external magnetic field thus obtaining a signal that depends on T_2 rather than T_2^*.

- After an initial 90° pulse has tipped the net magnetisation into the transverse plane the individual magnetic moments dephase due to spin-spin interactions (T_2) and due to inhomogeneities in B_0 (T_2^*).

- Spins experiencing a larger B_0 precess at a higher frequency and thus acquire a greater phase shift over a given time than spins experiencing a smaller B_0.

- At exactly half the echo time (TE/2) the refocusing pulse is applied. This flips the spins and effectively changes the sign of the relative phase change in the transverse plane. A large positive phase change becomes a large negative phase change and vice versa.

- Spins continue to gain or lose phase in the same direction as before by virtue of the magnetic field inhomogeneities. Thus, at time TE all of the spins come back into phase again.

- Throughout the spin echo sequence the irreversible spin-spin interactions continue so that the measured signal is still affected by T_2 dephasing.

- T_2 times are much longer than T_2^* and so spin echo signals tend to have higher SNR than gradient echo images, which are subject to T_2^* dephasing.

Spin Echo Formation

Signal (M_xy)

$90°$ $180°$

RF   t

TE/2   TE/2

G_{FE}

Spin echo

Signal
(M_xy)

T_2 relaxation curve

time

T_2^* relaxation curve

Signal rephased by combination
of 2nd magnetic field gradient
and $180°$ RF pulse
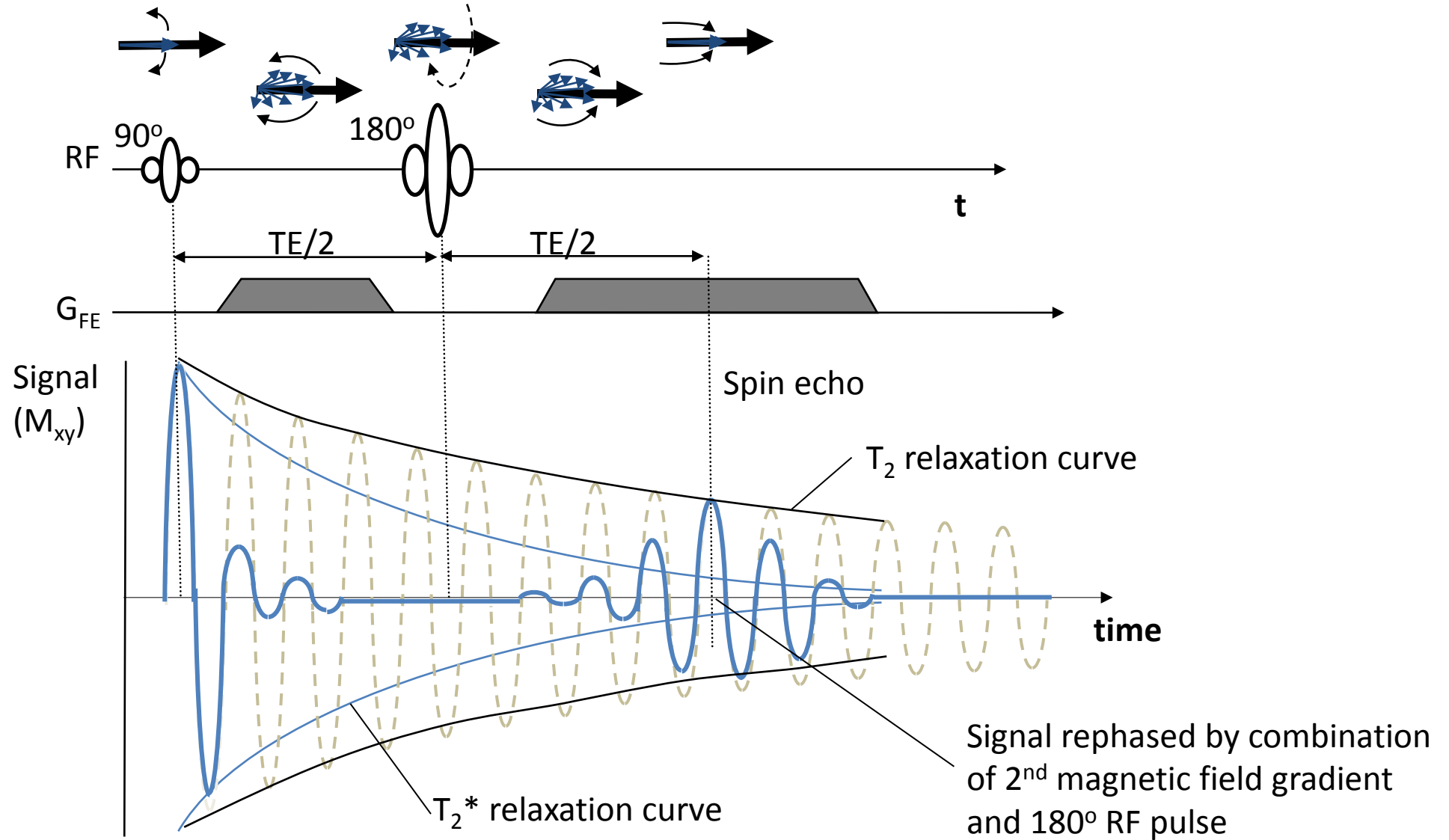
SE Pulse Sequence Diagram
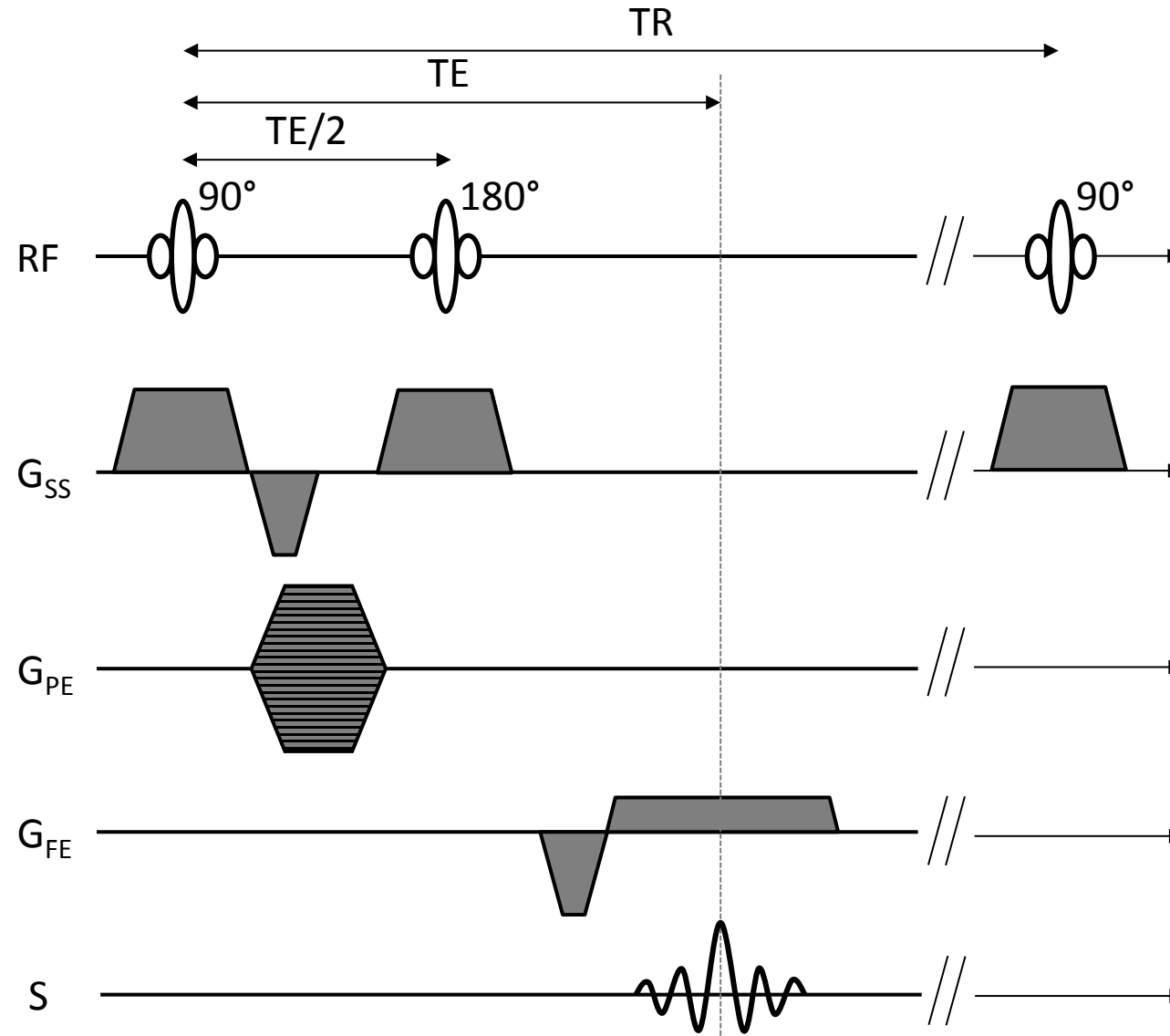
Spin Wash-out

- Mobile spins (e.g. blood) flowing through the imaging plane may not experience the effect of both slice selective RF pulses in the spin echo pulse sequence.

- Spins excited by the slice selective 90° pulse used in SE sequences can flow out of the imaging slice before the slice selective 180° refocusing pulse is applied.

- These spins are replaced by fresh unexcited spins from which no signal is measurable.

- Therefore flowing blood typically appears dark in SE images. The same mechanism affects Fast or Turbo Spin Echo sequences.

- In practice not all blood will necessarily wash out of the image slice, resulting in only partial blood signal suppression.

90° Slice Selective RF Pulse	Delay (duration TE/2)	180° Slice Selective RF Pulse
Spins in both blood and stationary tissue within the slice are excited	Excited spins in blood leave slice between RF pulses	In the image slice excited spins remain only in stationary tissue – no signal from blood
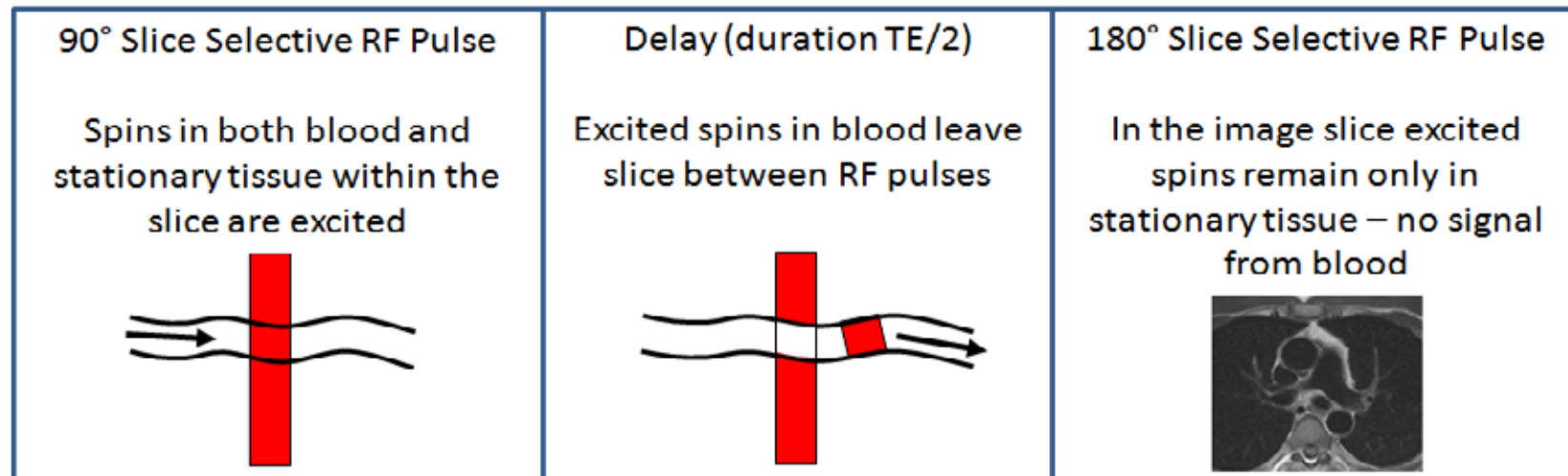
Conventional Spin Echo

- In conventional spin echo a single line of k-space is acquired each R-R interval.
- Each line employs a 90° pulse, to tip magnetisation into the transverse plane, and a 180° pulse to refocus dephased spins at TE and obtain an echo.

Fast / Turbo Spin Echo (FSE / TSE)

- After the initial first line of k-space is acquired the transverse magnetisation is repeatedly refocused using consecutive 180° pulses.

- A phase encoding gradient with a different amplitude is acquired before each echo.

- Equal and opposite rewinder gradients are applied after each echo to undo dephasing due to the phase encoding gradient and maintain the signal.

- The number of lines acquired after each 90° pulse is called the echo train length (ETL).

- The scan time for fast spin echo is reduced by a factor of ETL over conventional spin echo.

- T_2 decay persists throughout a given echo train so that each line has a different signal strength and T_2 contrast.

- As the dominant source of image contrast is the central line of k-space the effective TE (TE_{eff}) is determined by the echo time of the central k-space line.

Conventional Spin Echo

ECG

R-R

One line per
R-R interval

90°   180°

Scan time   n

Scan time = N_{PE} x TR

Fast/Turbo Spin Echo

ECG

R-R

Echo Train Length (e.g. ETL=4) lines
per R-R interval

Effective TE

Scan time   n

Scan time = (N_{PE} / ETL) x TR
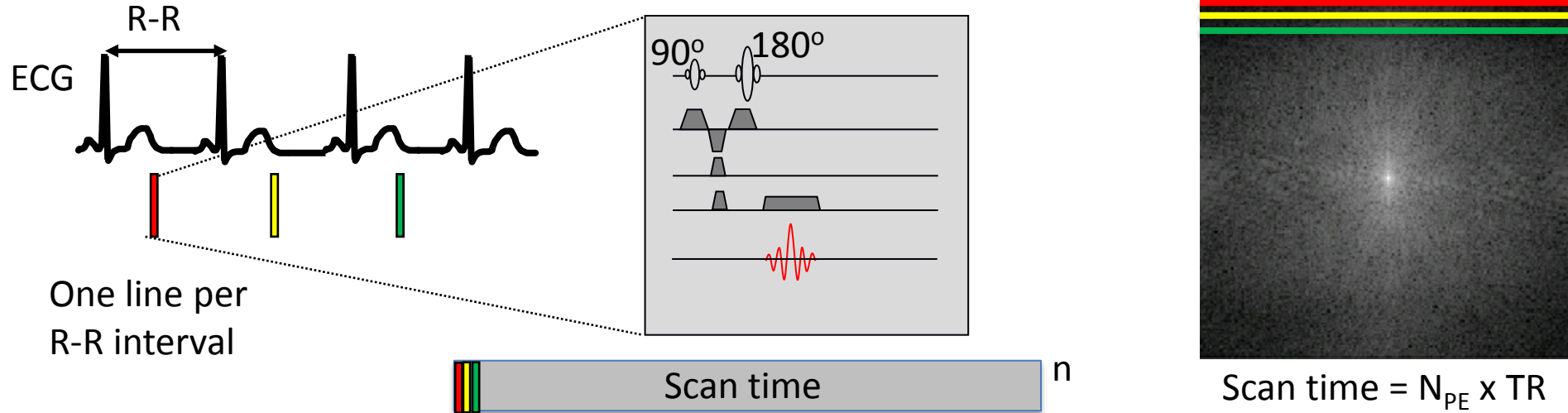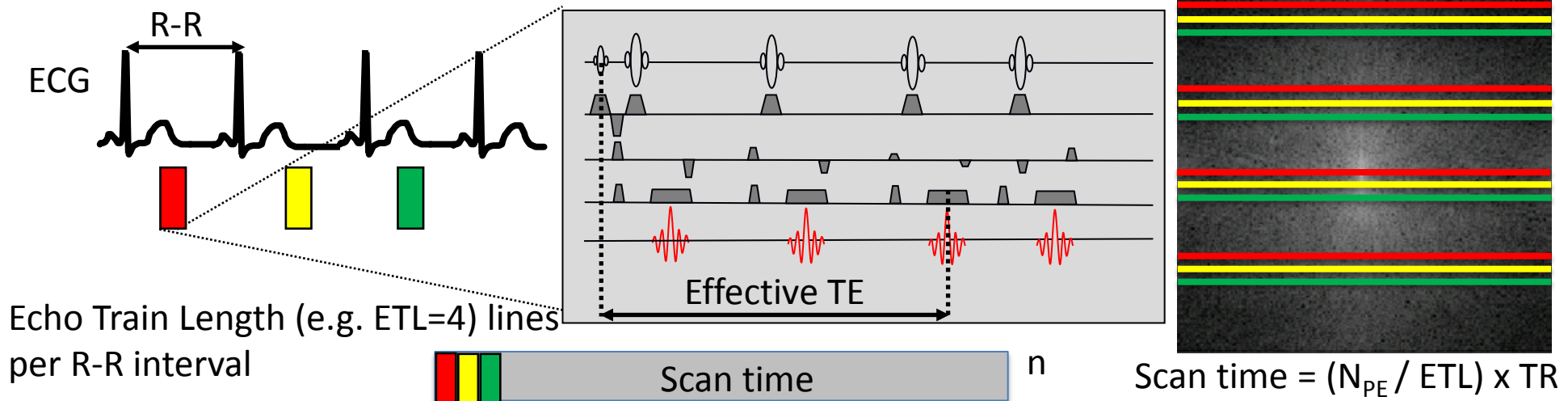
Double Inversion Recovery (IR) SE

- Black-blood contrast on spin echo images is unreliable, relying on a sufficiently high flow rate for spins to move out of the imaging slice within the TE/2 interval.

- Double IR addresses this by using two inversion pulses. The technique both nulls the signal from blood flowing in to the slice and increases the time interval within which blood can flow out of the slice.

- Two consecutive inversion pulses are applied in succession. The first pulse is non-selective and inverts all spins in the system, within and outside of the imaging slice.

- The second, slice-selective pulse re-inverts all the spins within the slice so that spins out side of the slice are inverted whilst spins within the slice are restored back to their original state.

- The image data acquisition is then performed after an inversion time (TI) chosen so that the longitudinal magnetisation of inverted blood inverted is nulled at the imaging time.

- By the time of image data acquisition most of the blood in the imaging slice will have been replaced by inverted blood from outside the slice.

- As the delay is substantially longer than the echo time, replacement of blood in the imaging slice will be more complete, even at low flows. The effects of double-IR preparation and spin-washout thus combine to improve blood suppression.

IR Based Nulling And Spin Wash-out

Inverted

Restored to original M_z

No signal

Non-selective 180°

Slice-selective 180°

RF

M_z

TI_{blood}

Image Data Acquisition

Stationary tissue has strong signal

Blood flowing in to slice is nulled
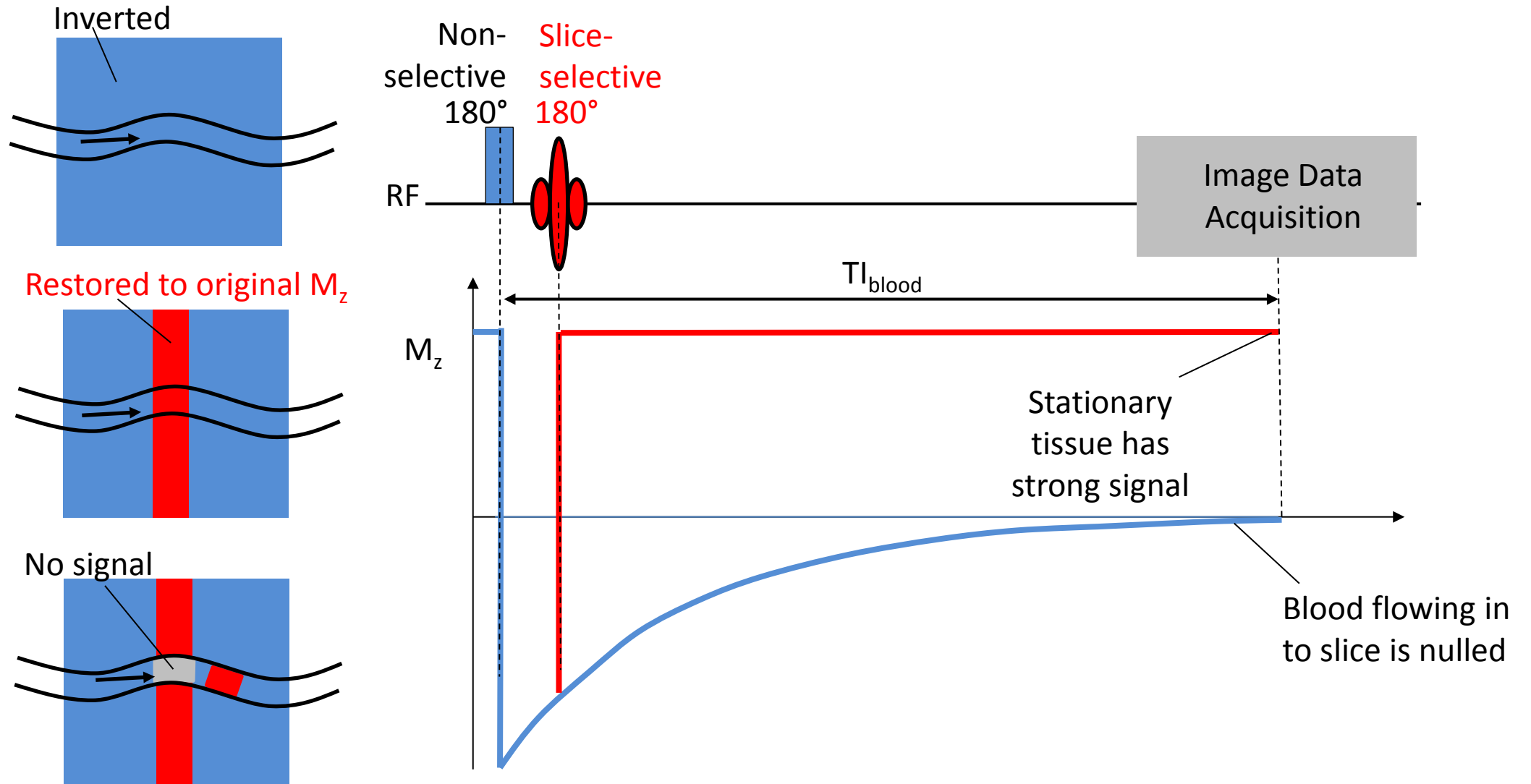
Triple IR SE – Blood And Fat Suppression

- In some cases it is necessary to suppress signal from both blood and fat (e.g. oedema imaging). A third slice selective inversion pulse between the double IR preparation and the image data acquisition can be used to achieve this.

- The inversion time (TI_{fat}) of the third inversion pulse is selected so that the longitudinal magnetisation of fat is nulled at the time of data acquisition.

- This techniques is often used with a long TE for T_2 weighting to visualise oedema (as fluid, which has both long T_1 and T_2, will be bright).

Non-selective 180° Slice-selective 180° Slice-selective 180°

RF

Image Data Acquisition

TI_{blood}

M_z

Fat is nulled

Blood flowing in to slice is nulled

TI_{fat}

Stationary non-fatty tissue has strong signal
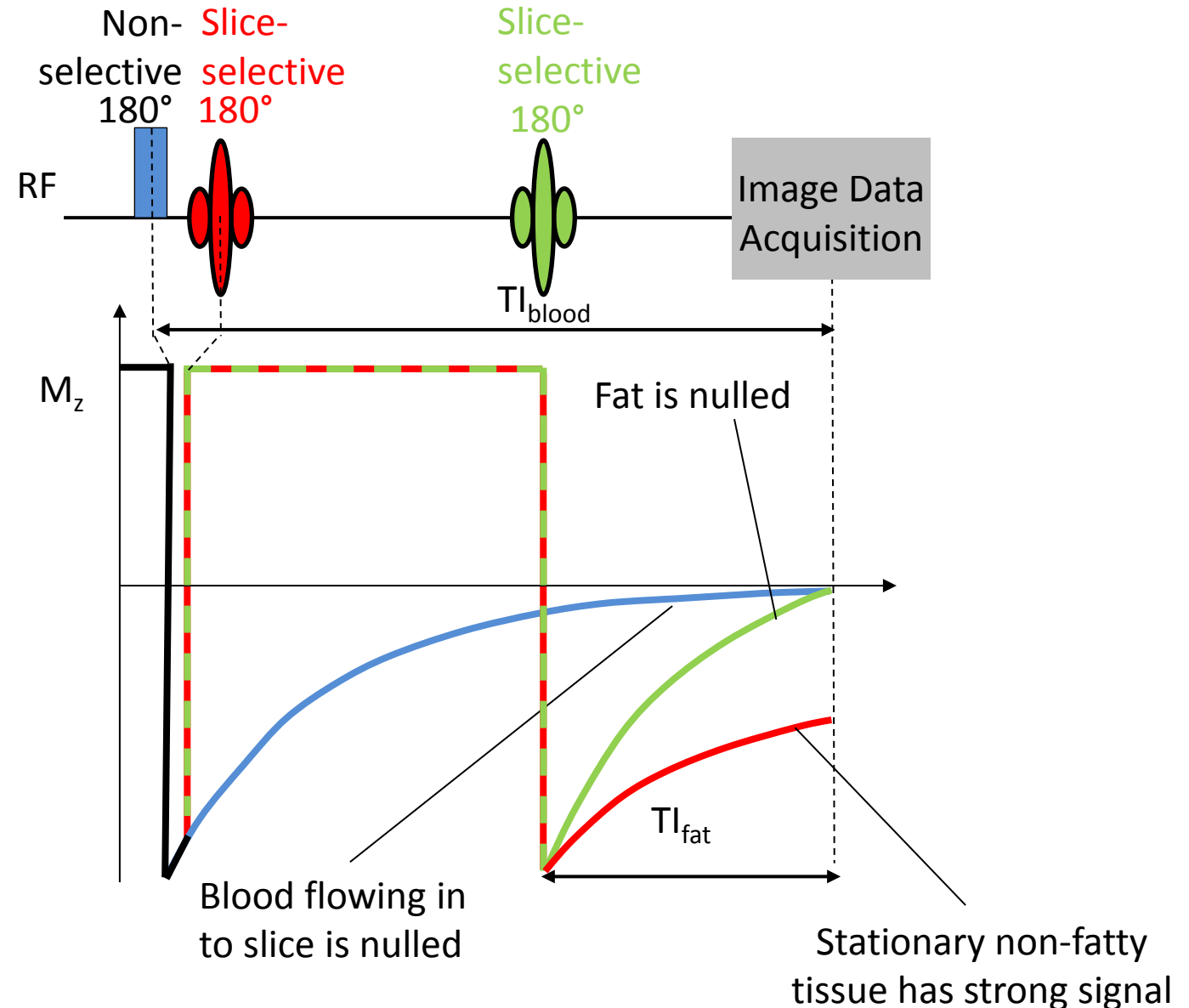
Background

- In MRI the signal arises from hydrogen nuclei. The dominant sources of signal are from water and fat. Due to the different chemical environment of hydrogen nuclei in fat the spins precess at a slightly slower frequency than those in water. This chemical shift is about 3.5 parts per million (ppm), or around 220 Hz at 1.5T and 440 Hz at 3T.

- It may be desirable to suppress signal from fat to improve visualisation of non-fatty tissues. Alternatively it may be useful to compare images with and without fat suppression to distinguish fatty lesions (e.g. lipomas) from other tissue types.

- Several methods have been developed to suppress signal from fat, including the triple-IR prepared SE sequence described previously. These exploit either the short T_1 of fat or its chemical shift (or in some cases both).
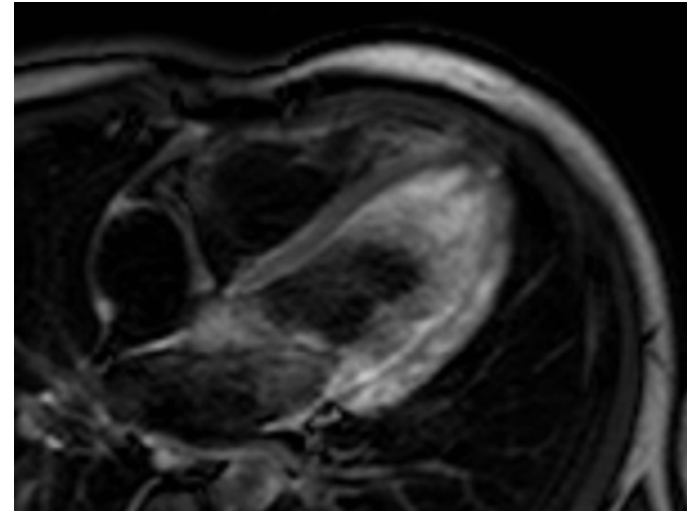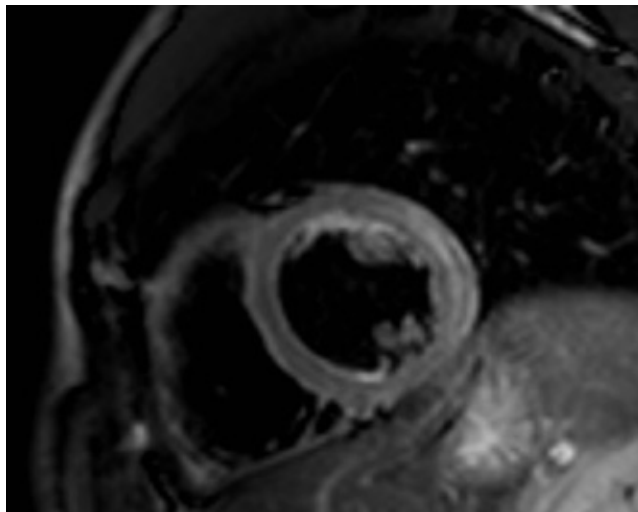
Short Inversion Time Inversion Recovery - STIR

- STIR uses a 180° inversion pulse followed by an inversion time (TI) before reading out the signal.

- Following inversion longitudinal magnetisation will recover to equilibrium, passing through zero (the null point).

- Fat has shorter T_1 than other tissues and fluids (in the absence of contrast agents) and so passes through the null point first.

- Fat suppression can be achieved by acquiring image data at the null point of fat, at which time magnetisation from other tissues will still be inverted and recovering.

- Signal intensity is proportional to the longitudinal magnetisation magnitude immediately prior to data acquisition. Consequently, unlike normal T_1 weighting, tissues with the longest T_1 will generate the highest signal.

- STIR can be combined with black-blood preparation to suppress blood and fat signal (triple IR imaging). This can be particularly useful for imaging oedema when combined with a long TR (2-3 RR intervals) and long TE to achieve T_2 weighting.

STIR – Fat Nulling

RF

180°

TI_{fat}

Data Acquisition

TI = fat null point

Negative magnetisation
Longer T1 = greater signal

Fat (Short T1)
Medium T1
Long T1
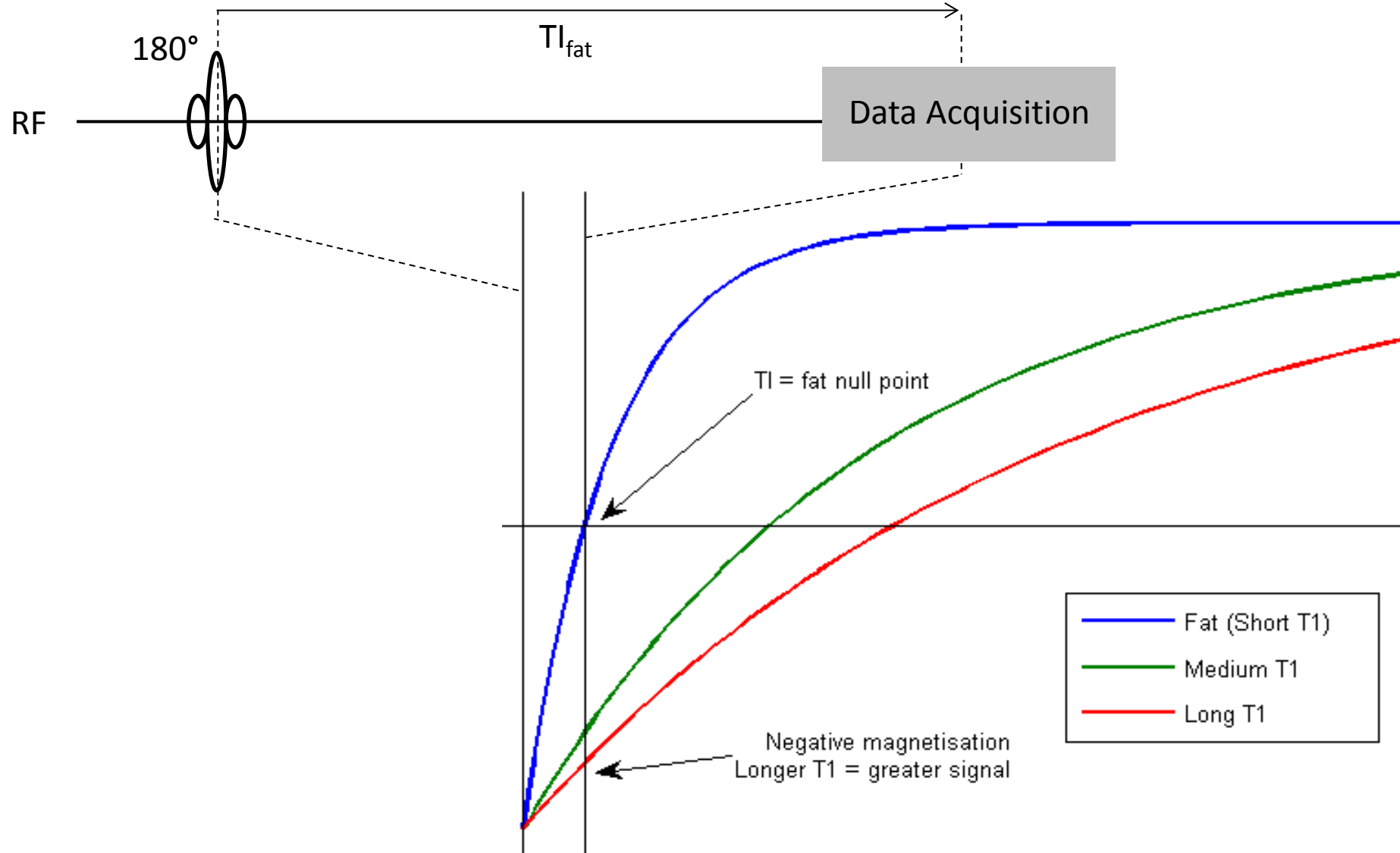
## Chemical Shift Selective (CHESS) Magnetisation Preparation

- As well as suppressing fat, signal from longer T_1 tissues and fluids is also partially reduced in STIR, modifying T_1 weighting and reducing SNR.

- This can be avoided by exploiting the difference in resonant frequencies between fat and water rather than the difference in T_1.

- By applying a spectrally selective preparation pulse, the longitudinal magnetisation of fat is altered whilst leaving the water signal unperturbed. Fat signals will be suppressed in images acquired after this preparation due to the reduced longitudinal magnetisation available.

- Chemical Shift Selective (CHESS) preparation is highly dependent on magnetic field inhomogeneity and good shimming is required. While good shimming can be harder to achieve at higher field strengths the separation of the fat and water peaks also increases making the technique easier.

- A saturation pulse followed immediately by image data acquisition can be performed ('fat sat'), although this will allow partial recovery of magnetisation in fat prior to acquisition of the central k-space data. Alternatively an inversion preparation and appropriate delay can be used to avoid this problem. Commonly an intermediate flip angle (around 120°) is used so that the magnetisation is only partially inverted (Spectral Presaturation with Inversion Recovery, SPIR) and a shorter delay can be used.

CHESS Fat Suppression
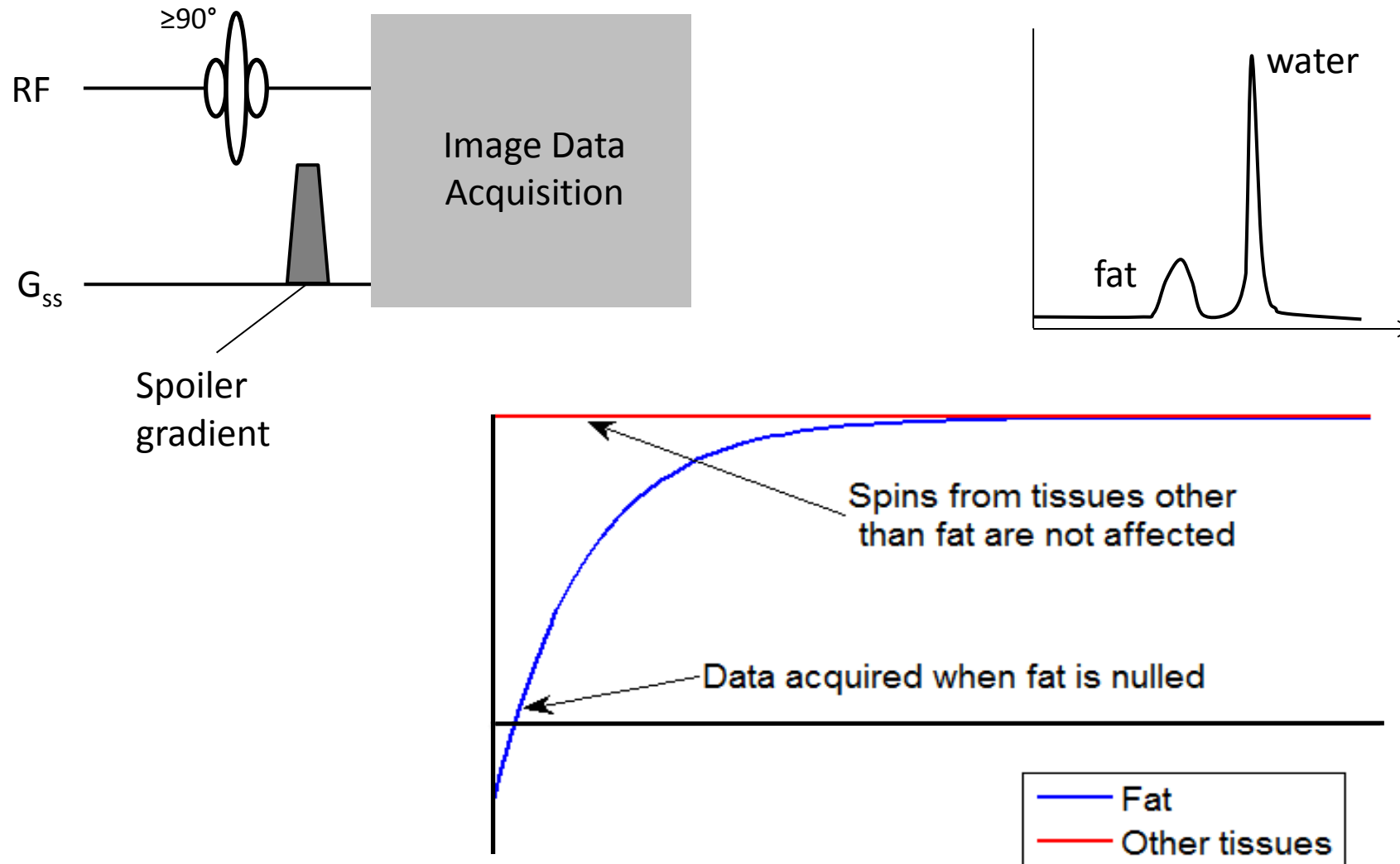
Background

- Velocity information can be encoded in the phase of the transverse magnetisation at the time of data acquisition. This can be seen by viewing the phase map that is calculated as part of the image reconstruction process. This is normally discarded and only the signal magnitude used to generate images.

- In gradient echo imaging bipolar gradient pulses are commonly employed so that the net phase difference of stationary tissues is zero at the centre of the echo. The effect of the positive and negative gradients will not cancel out exactly for moving spins, so a phase difference will accrue.

- For a bi-polar pair of rectangular gradients and constant flow the phase difference is proportional to velocity. Flow sensitivity can be increased by increasing the strength, duration or separation of the bipolar gradients.

- The phase image (top) shows flow through the aortic valve. Bright voxels show flow out of, and dark voxels flow into, the ventricle. Stationary tissue appears mid-grey. The lower image shows the corresponding anatomy (signal magnitude).
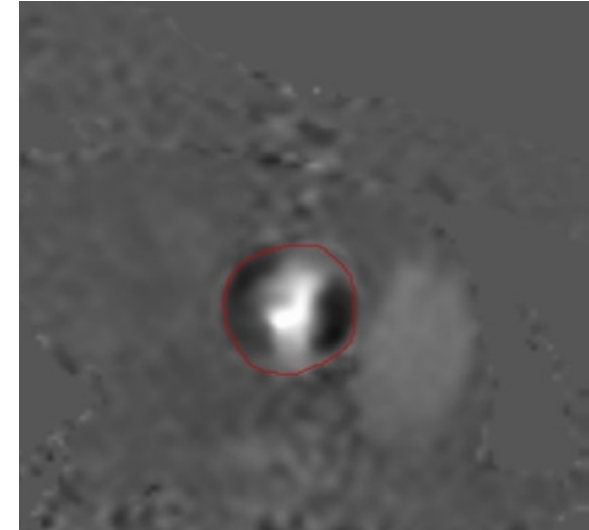
Velocity Encoding

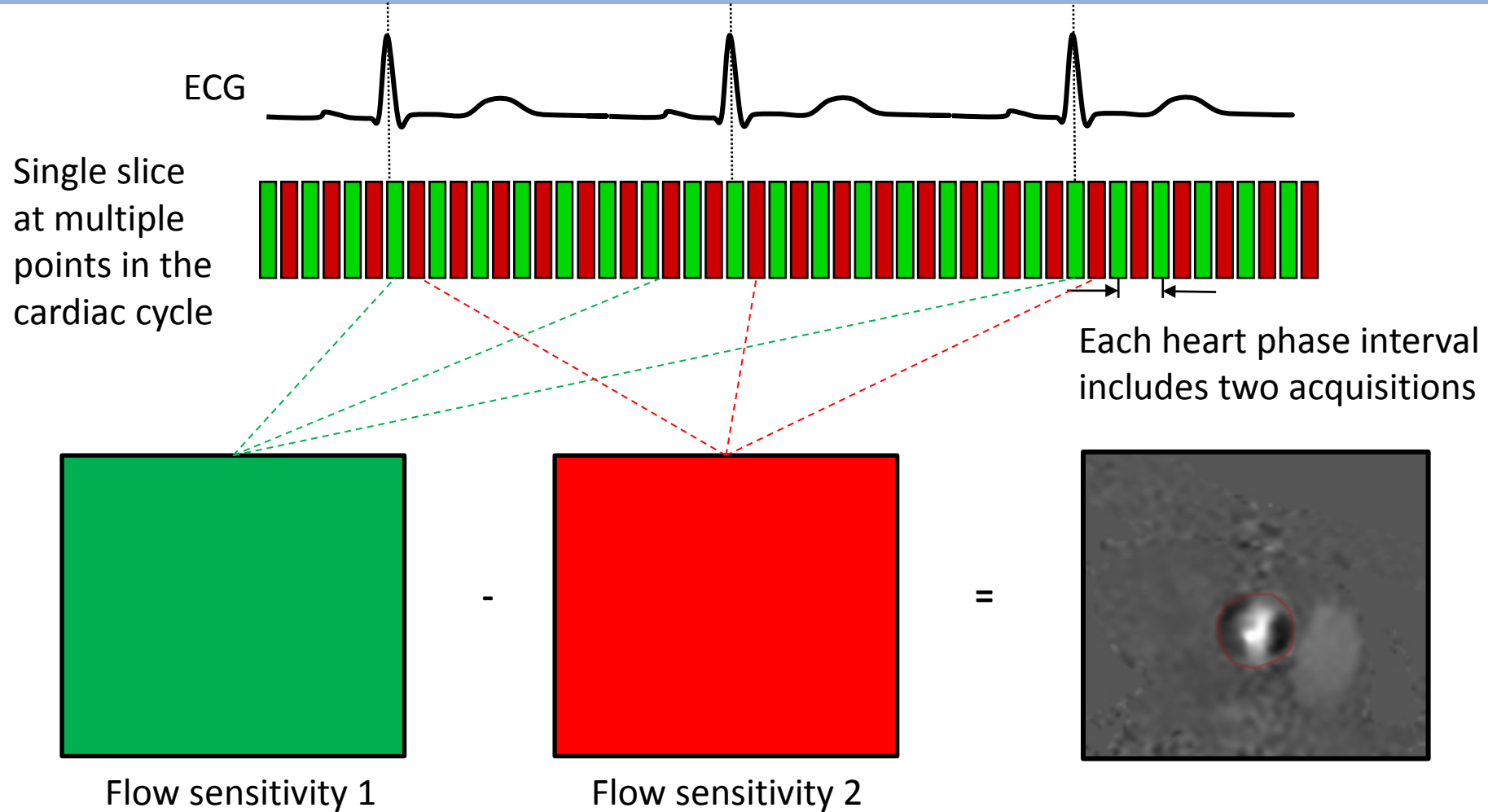- Velocity is encoded in the phase (ϕ) by a bipolar gradient pair. The net effect of a bipolar gradient pair on the phase of stationary spins is zero, but for moving spins the effect of the two pulses is not balanced.

For stationary spins no net change of phase

Moving spins acquire a phase change

*flow in phase contrast velocity encoding direction
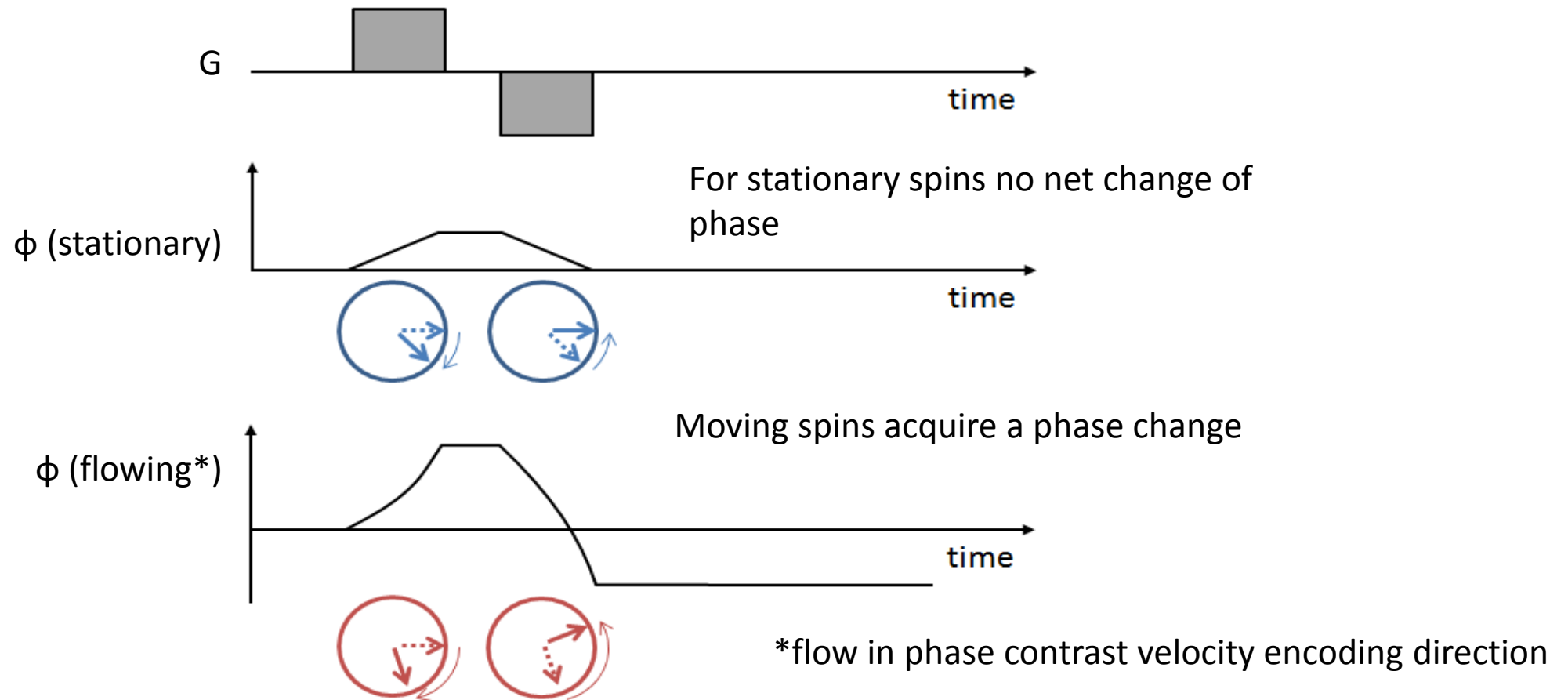
Acquisition

- Velocity encoded images are typically acquired using a cine imaging strategy, so that changes in velocity over the cardiac cycle can be seen.

- Phase changes can also be induced by motion in directions other than the velocity encoding direction and due to magnetic field inhomogeneities.

- In order to isolate phase changes due to motion along the desired direction two consecutive acquisitions are acquired, with two different flow sensitivities, for each phase of the cardiac cycle.

- Magnetic field inhomogeneities and motion in the non velocity encoding direction should remain constant for the two acquisitions. Therefore, subtraction of the two phase maps should result in an image where phase change is due only to motion in the velocity encoding direction.

- The maximum measurable velocity range (known as the VENC) is determined by the difference in flow sensitivities between these two acquisitions.

Acquisition

ECG

Single slice at multiple points in the cardiac cycle

Each heart phase interval includes two acquisitions

Flow sensitivity 1   Flow sensitivity 2   =

Each frame is acquired with two flow sensitivities which are subtracted (to correct for phase offsets due to other effects) to create the final phase map

VENC and Aliasing

- If the VENC is set appropriately then the range of phase shifts will fall between -180º and +180º, corresponding to positive and negative motion along the velocity encoding direction. However, if the VENC is set to low then phase shifts outside this range are possible. A phase shift of +190° is indistinguishable from a phase shift of -170°. Velocities above the VENC will therefore be aliased (i.e. positive velocities greater than the VENC will look like negative flow).
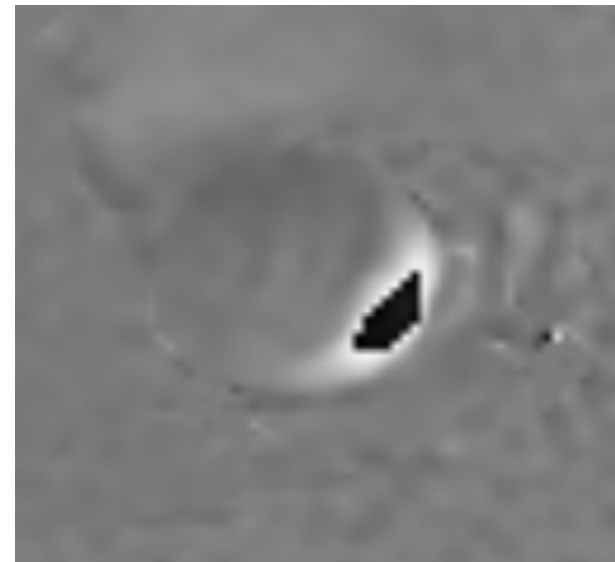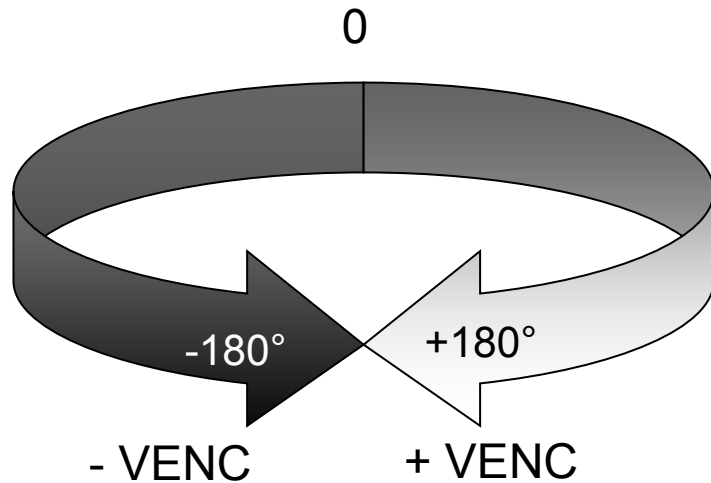
Phase image acquired with 250cm/s VENC. Flows that exceed that velocity are aliased

Background

- The use of a tagging scheme (e.g. Spatial Modulation of Magnetisation, SPAMM) produces a geometric pattern of magnetisation in tissue. This modifies the appearance of images acquired for a short time after the tagging scheme application.

- As the magnetisation pattern moves with the tissue this can help visualise and quantify myocardial motion.

- By tracking the motion of tagging features quantitative metrics, such as myocardial strain, can be calculated.

- Tagging can allow assessment of circumferential motion that is not apparent in conventional cine imaging.
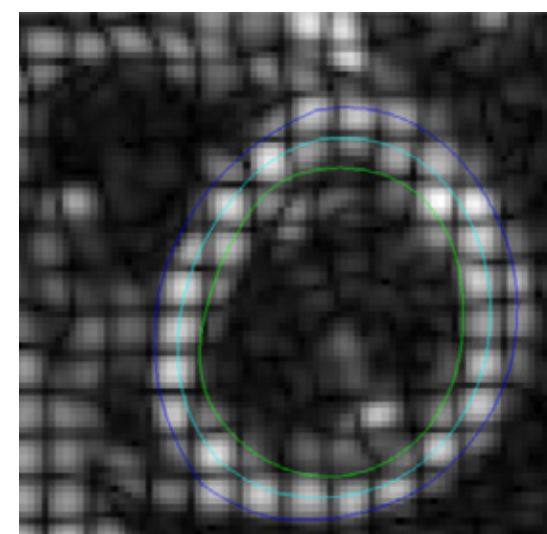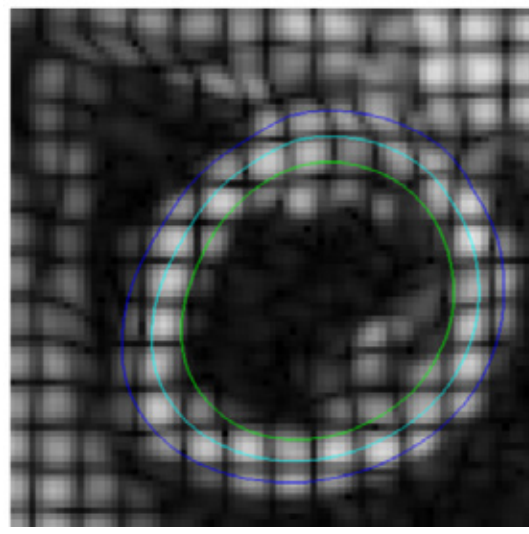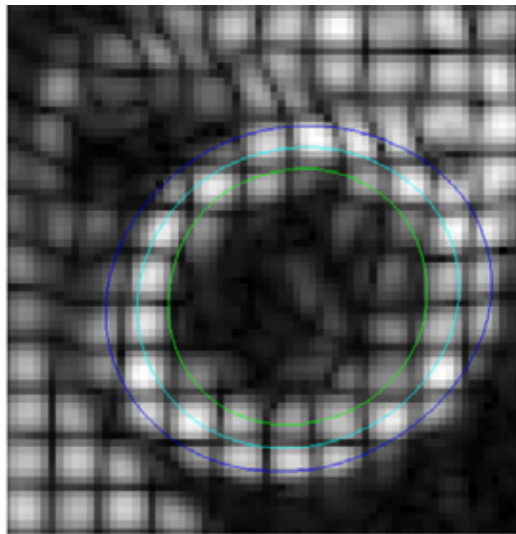
SPAMM – Pulse Sequence Diagram

- The SPAMM preparation scheme consists of a composite pulse consisting of multiple RF pulses and modulating gradients.
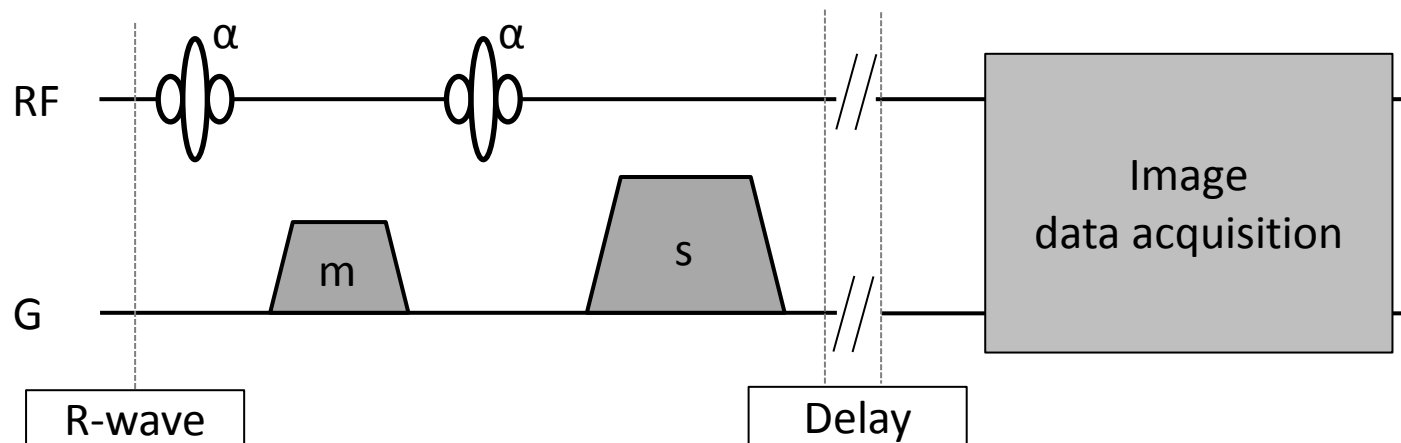
- In the basic SPAMM implementation shown above the first RF pulse rotates magnetisation from the longitudinal axis. The modulating gradient (m) then imparts periodic phase variation across the image plane. The second RF pulse rotates magnetisation further in places where the net phase is zero, and reverses the effect of the first pulse for spins with opposing phase. Typically a total flip angle of 90° is used so that magnetisation in some regions becomes saturated, although this may be achieved by a larger number of RF pulses and modulating gradients.

- A spoiler gradient(s) removes transverse magnetisation prior to standard cine image acquisition throughout the cardiac cycle.

Magnetisation Modulation – 1-2-1 SPAMM
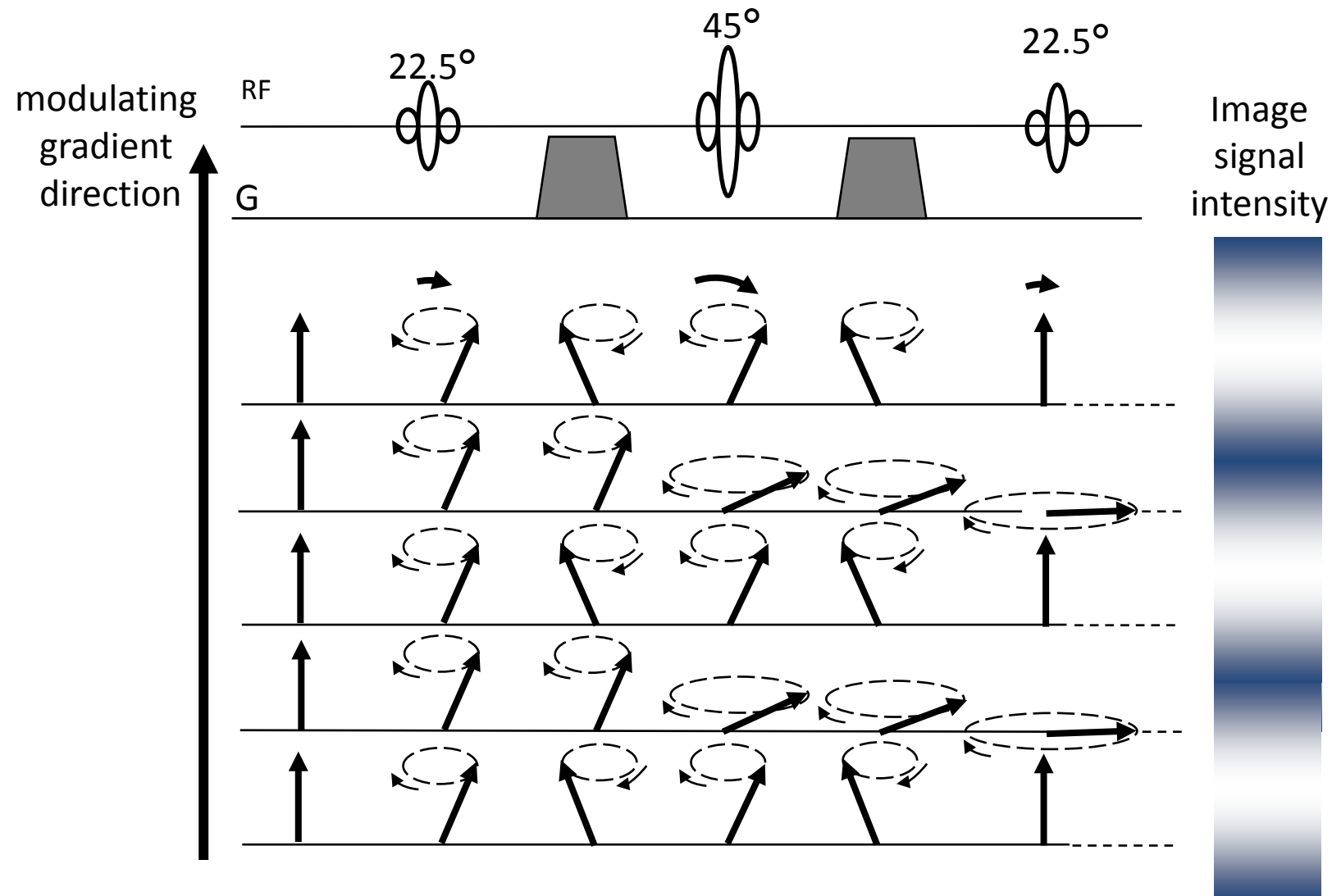
A 1-2-1 flip angle scheme is shown here. Higher numbers of pulses with flip angles following a binomial pattern (e.g. 1-3-3-1) can also be used to achieve sharper tag lines

Tag Contrast

- Tag contrast decreases with time due to longitudinal recovery (T_1) of the magnetisation in tagged regions. This leads to decreased tag line conspicuity and increased visibility of underlying anatomy throughout the cardiac cycle.
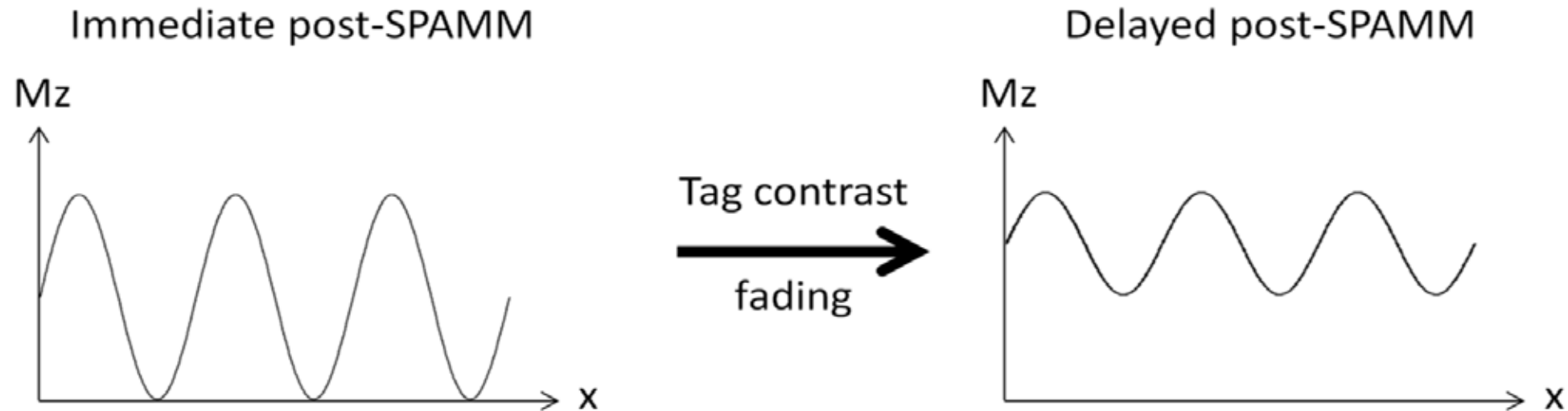
- Tags will be more persistent the longer the T_1 of the tissue. Consequently tag persistence is better at higher field strengths due to the longer T_1.

- Tag fading will be much more rapid if a tagging sequence is performed after contrast agent administration.

Tagging Patterns

- The basic SPAMM scheme imparts a parallel lines pattern with the tag line spacing controlled by the modulating gradient strength (a stronger gradient leads to closer tag line spacing).

- A second preparation phase can be applied with a perpendicular modulating gradient to achieve a grid pattern.

- Alternative tagging patterns (e.g. radial) have also been used.

- In the images below the tag lines are distorted in the myocardium due to motion between tagging and image data acquisition, but have maintained their original pattern in neighbouring stationary tissue.
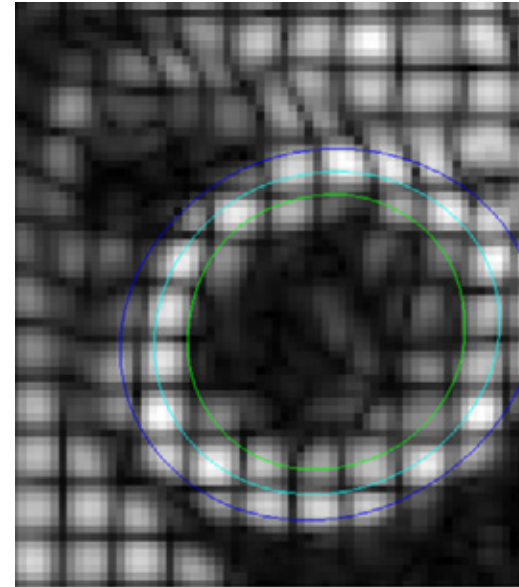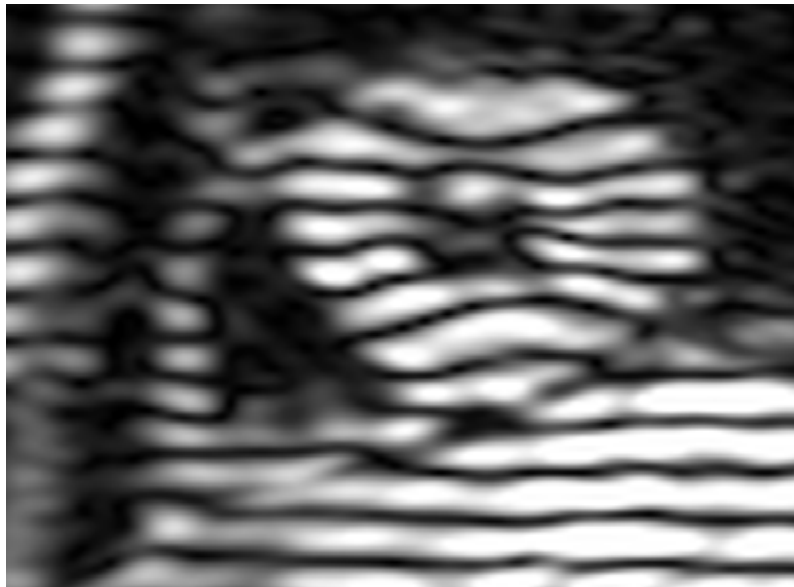
Complementary Image Subtraction

- In Complementary Spatial Modulation of Magnetisation (CSPAMM) two images with opposing tag patterns are acquired.

- If the first image uses a 90°/90° RF pulse pair the second will use a 90°/-90° pair. Subtracting the images yields a tagged image with:
    - Consistent nulling of tagged myocardium throughout the cardiac cycle.
    - Increased SNR and tag contrast compared to SPAMM

- However:
    - The tag contrast still fades with time.
    - Acquisition of each cardiac phase takes twice as long.

- A tag-free image can also be reconstructed by adding the complementary images.

Ramped Flip Angle

- Tipping magnetisation into the transverse plane during image data acquisition accelerates fading of tag contrast.

- Increasing the readout flip angle throughout the cardiac cycle can improve tag persistence. The optimal flip angle ramp profile depends on the imaging sequence used.

Consistent Nulling

- Even after a delay full nulling of signal in the tagged regions can be achieved with CSPAMM by subtracting images with out-of-phase tagging patterns.

- Additionally an untagged image can be created by adding the two tagged images.
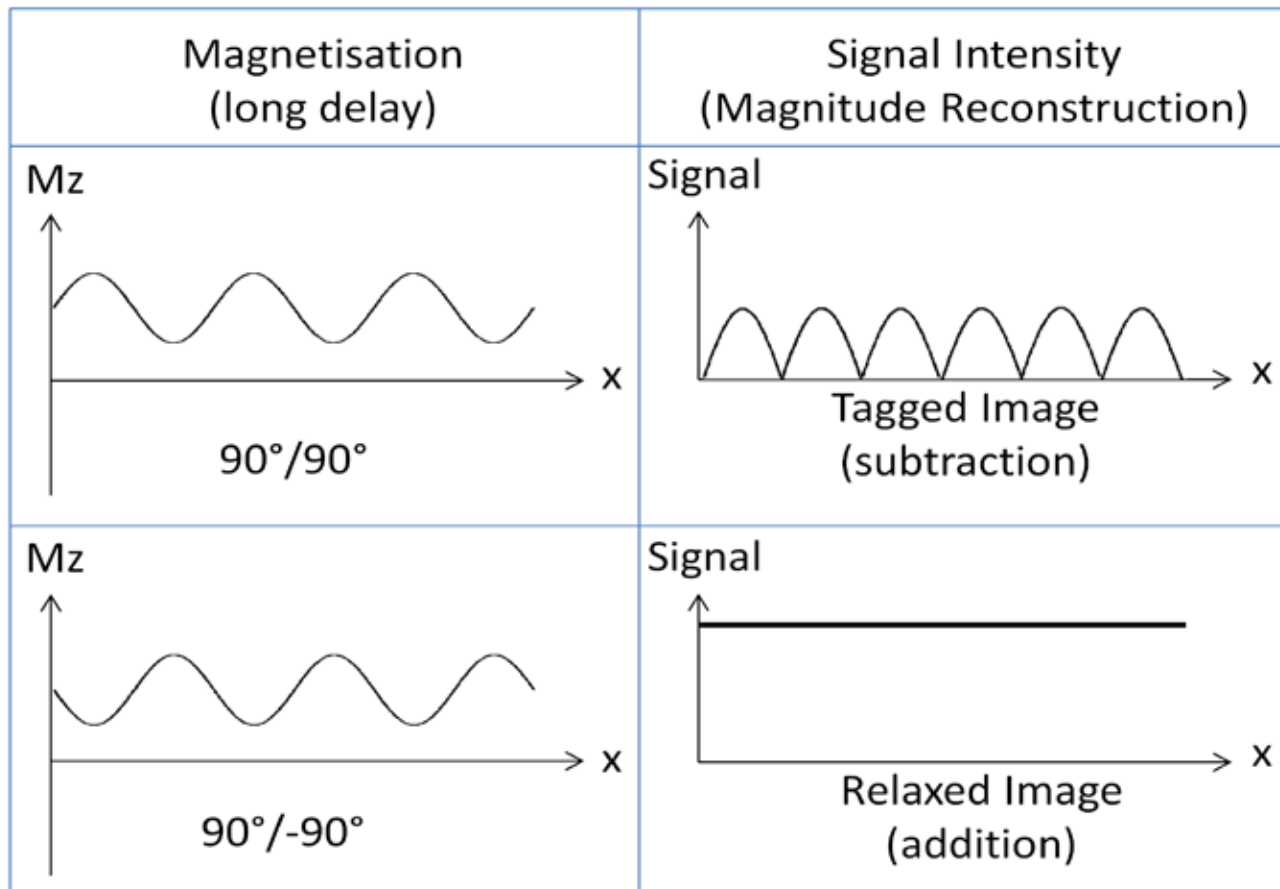
Late Gadolinium Enhancement

- Viability, or late gadolinium enhancement (LGE), imaging exploits the fact that the distribution volume accessible to the contrast agent is larger in infarcted tissue as cell membrane integrity is disrupted, allowing the agent to enter the intracellular space.

- As a result scar tissue appears bright on T_1 weighted images taken at an appropriate delay post-contrast administration (~15mins).

- Viability imaging is performed using a non-selective 180° inversion preparation pulse. Image data is acquired when signal from normal myocardium is nulled to optimise the contrast between the bright hyper-enhancing scar tissue and the normal myocardium. As the T_1 of normal myocardium varies from study-to-study (due to variation in contrast dose, protocol and physiological parameters) a TI-scout, Look-Locker sequence with multiple TI values is used to visually identify the optimal TI value for the LGE sequence.

M_z

TI_{myo}

Scar tissue

blood pool

myocardium

time after inversion pulse

Data acquired when myocardium is nulled
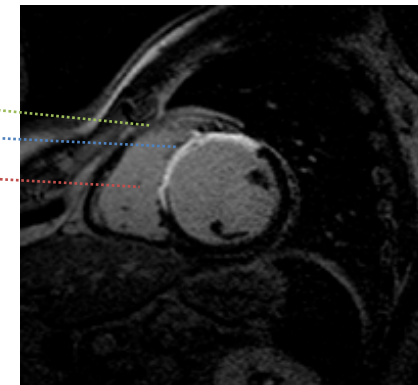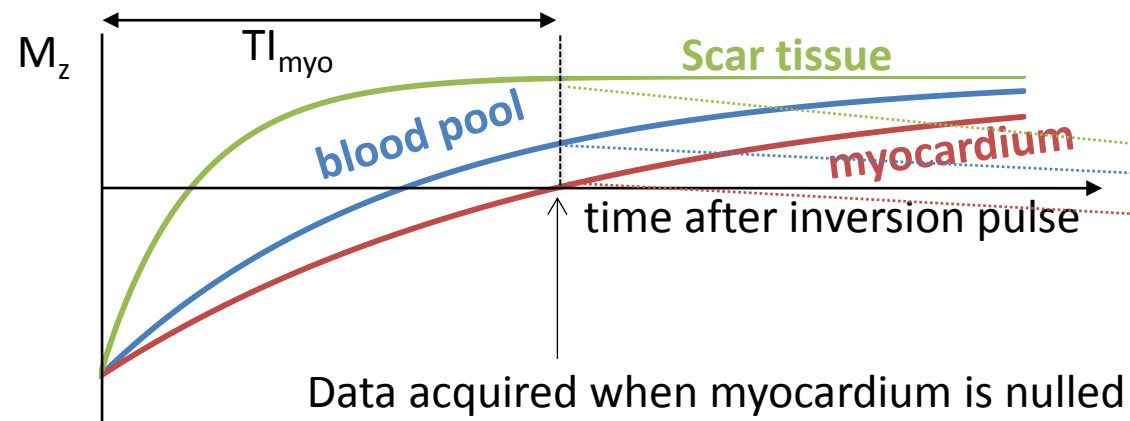
Signal from viable myocardium is nulled in the final image

LGE Acquisition

- Images must be T_1-weighted with a high spatial resolution and whole heart coverage (requiring multiple contiguous slices). Therefore a segmented read-out sequence is used to acquire the image data over multiple heart beats.

- Typically 180° pulses are separated by two RR intervals to allow sufficient recovery of M_z.

- A single slice is acquired per breath-hold, with whole heart coverage being achieved through multiple breath-holds. The achievable breath-hold time limits the achievable spatial resolution.

- TI_{myo} must be modified throughout the exam to accommodate the gradual washout out of contrast agent from the myocardium which leads to changing T_1.
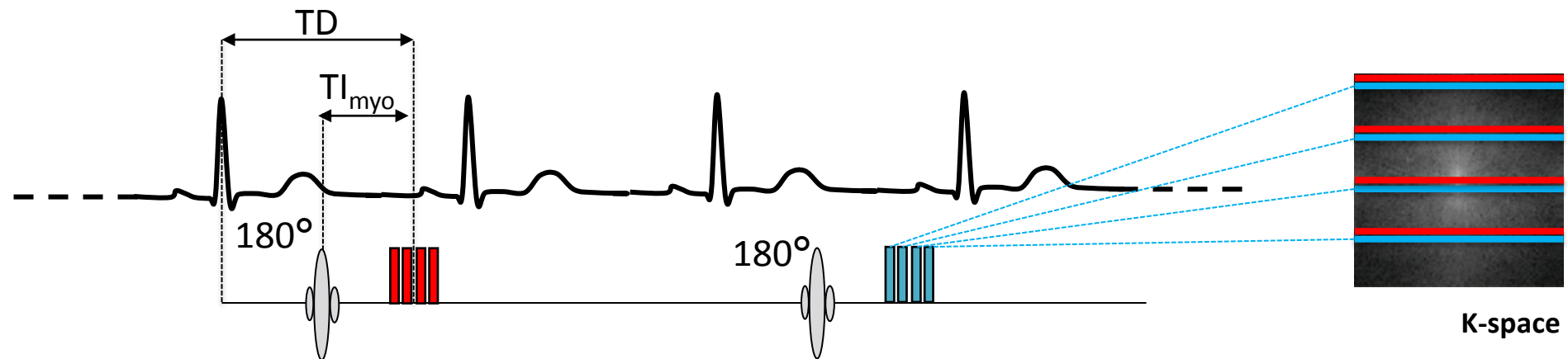
Segmented acquisition constructs k-space over multiple heart beats

Myocardial Perfusion Imaging Method

- The myocardial perfusion imaging acquisition is designed to display the passage of contrast agent over time through a fixed slice of myocardial tissue at a fixed point in the cardiac cycle. Images from successive heart beats are displayed as a movie, showing an increase in signal intensity during the first pass of contrast agent through the myocardium. Hypo-intense areas on the resulting movie indicate myocardial ischaemia.

- To visualise myocardial ischaemia imaging must be performed under stress conditions (typically pharmacologically induced). A rest scan may also be taken for comparison

- In order to visualise changes in signal due to contrast agent passing through the heart images must be acquired every RR-interval. In addition at least three slices must be acquired to obtain sufficient coverage of the heart. To achieve this a very fast read-out sequence (Fast/Turbo Gradient Echo, bSSFP or hybrid EPI) is used in addition to other acceleration techniques such as parallel imaging and partial Fourier.

- Images must be T_1-weighted to visualise the T_1 shortening effect of the contrast agent. This is achieved using a saturation prepared sequence.

Myocardial Perfusion Imaging Method

Contrast injection

RR-interval

ECG

TD_1

TD_2

TD_3

PPD PPD PPD

Saturation Prep Pulse

Slice 1 Slice 2 Slice 3

90° 90° 90°

Fast/Turbo GE, bSSFP or Hybrid/Segmented EPI imaging sequence

Basal Mid Apical
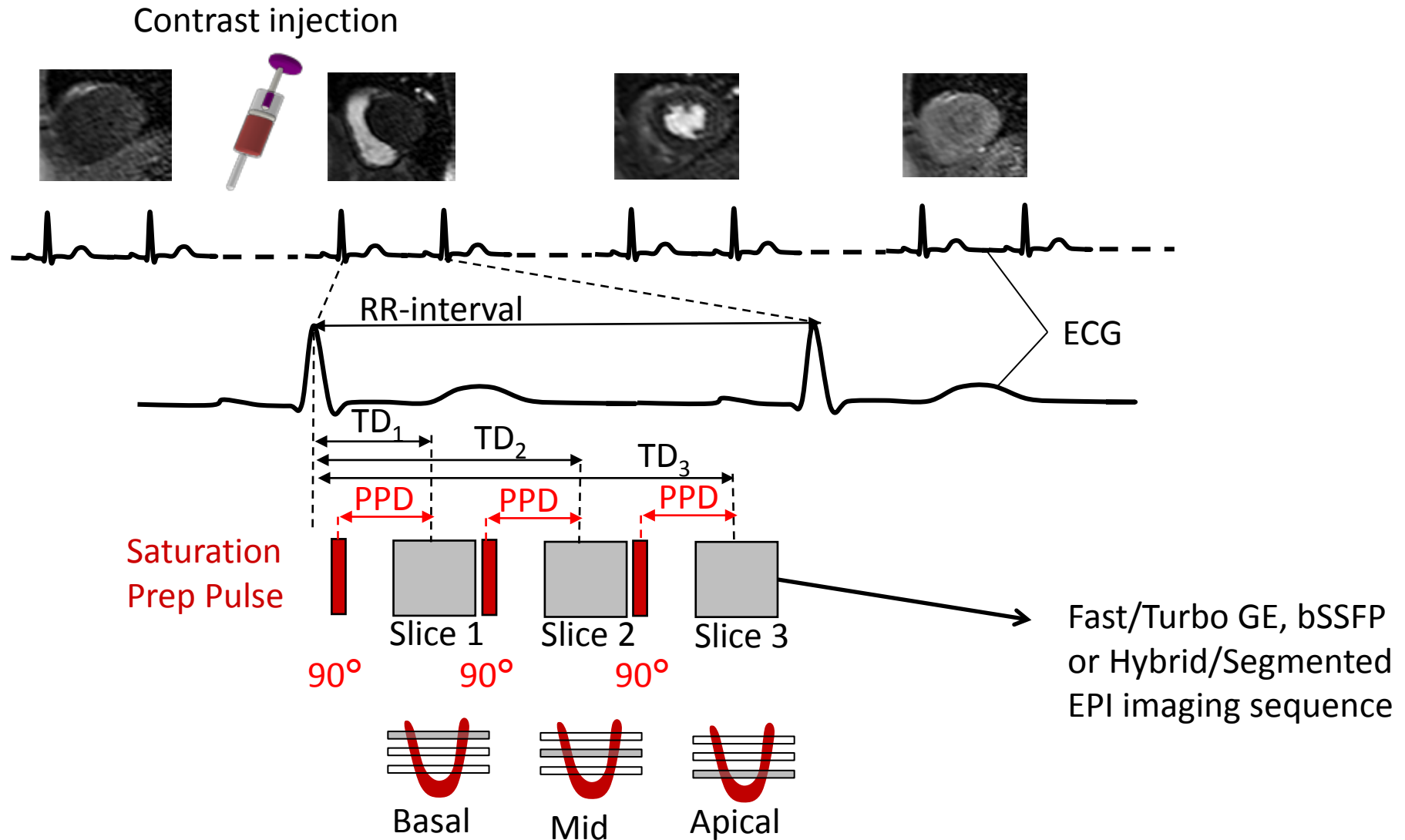
Myocardial Perfusion Imaging - Imaging with High Heart Rates

- As imaging is performed under stress the RR-interval can become too short to allow time for the pulse sequence and the scan time must be reduced by:
  - Shortening PPD – only possible if there is unused time between the saturation pulse and the read-out train. Reduces T_1-weighting in images.
  - Reducing image resolution – increases potential for missing subendocardial defects.
  - Reduce coverage – potential for not imaging regions of ischaemia.
  - Extend imaging over two RR-intervals – reduced temporal resolution.

Analysis

- Myocardial perfusion images can be assessed visually to identify regions of hypo-perfusion.

- Semi-quantitative metrics can be determined by assessing the variation of signal intensity of the myocardium over time.

- Quantitative measures of myocardial blood flow can be made by analysing the variation of the signal intensity of the myocardium and the left-ventricular blood pool. Due to the higher contrast agent concentrations that are encountered in the blood pool modifications to the image acquisition or contrast agent administration protocol may be required to be able to perform this accurately.

Relaxometry ('Mapping')

Background

- By adjusting scan parameters the amount of T_1 and T_2 (spin echo) or T_2* (gradient echo) can be controlled. However signal intensity is displayed on an arbitrary scale and is still influenced somewhat by all of the tissue relaxation time constants as well as the density of signal generating spins (the proton density).
- It is sometimes desirable to generate quantitative images, or maps, where the voxel intensity values directly represent the relaxation time constants.

General Methodology

- In general relaxometry methods rely on the acquisition of multiple images and subsequent post-processing following the steps below:

  - Acquire multiple images with different contrast weighting.

  - If required perform motion correction or image registration to account for movement between acquisition of the different images.

  - Fit signal intensities to a model describing the pulse sequence.

  - Extract relaxation times from the fit results to generate a parametric map.

T$_2$ Mapping Methodology

- SE images are acquired with varying TE (and so varying T$_2$ weighting)

TE = 15.4 ms   TE = 29.1 ms   TE = 42.8 ms   TE = 56.4 ms   TE = 70.1 ms   TE = 83.7 ms
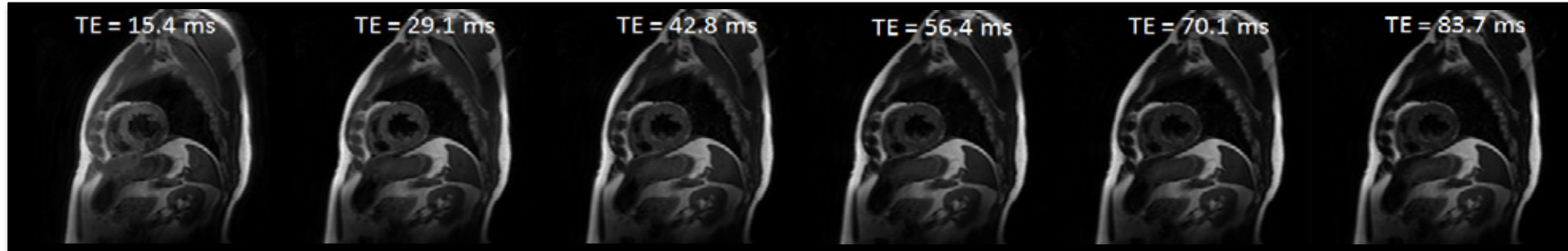
- An exponential decay model is fitted to the acquired signal intensities for each voxel.

$S=S_0\exp(-TE/T_2)$
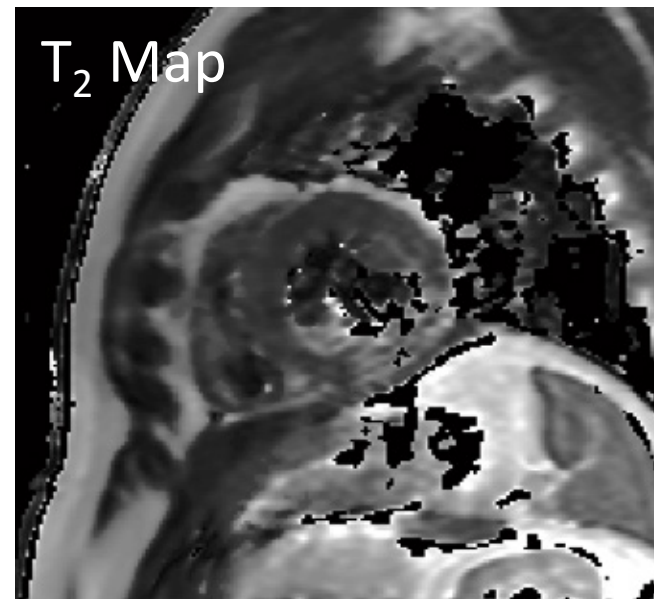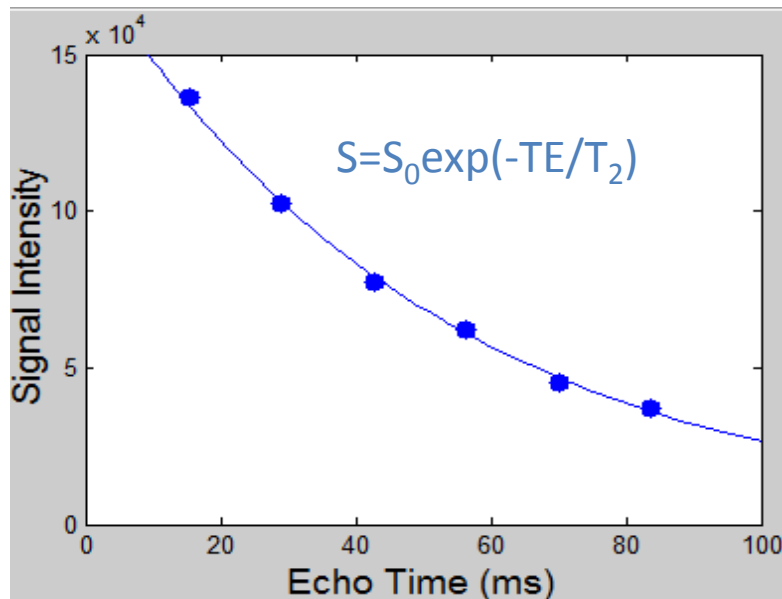
T$_2$ Map

T$_2$ & T$_2$* Mapping

Uses

- T$_2$ mapping may be useful for identifying myocardial oedema and haemorrhage, currently in a research setting.
- T$_2$* maps can be generated similarly to T$_2$ mapping but using GE (rather than SE) images with varying TE. T$_2$* mapping may help visualise processes such as myocardial iron deposition.
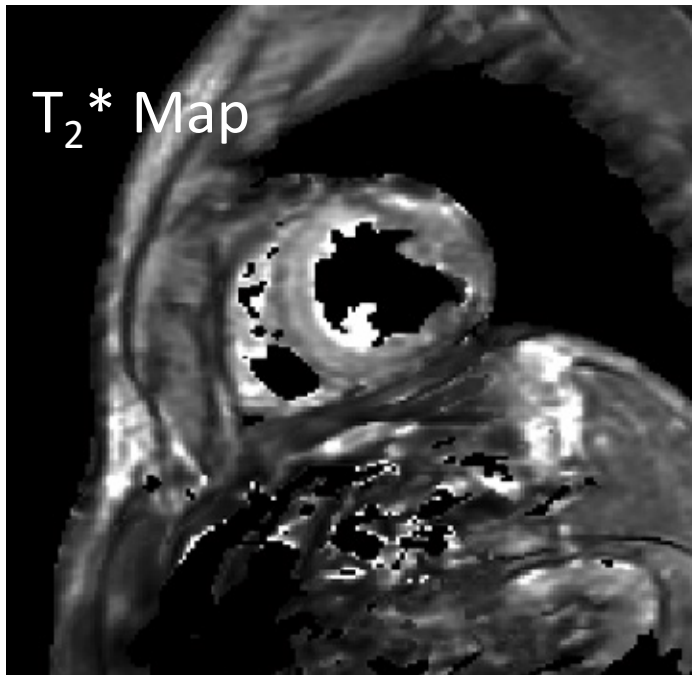
T$_2$* Map

T$_2$ and T$_2$* Mapping Considerations

- TE selection
  - A suitable range of TE values to accurately and precisely characterise the longitudinal magnetisation decay.
  - Use of long TE images with low signal can lead to bias in the estimated T$_2$ or T$_2$* values.
- Susceptibility artefact
  - Susceptibility differences close to veins can lead to artefactual T$_2$ or T$_2$* values. This commonly occurs in the infero-lateral myocardium, close to the posterior vein.

Original Basis of Look-Locker Method

- The Look-Locker method was originally developed to measure T_1 values in MR spectroscopy.
- Data is acquired continuously after an inversion pulse to sample longitudinal recovery. The collected data is fitted to an exponential recovery equation.

Signal Model

- For undisturbed recovery following an inversion pulse the following signal model applies:
  - $S = S_0 (1 - 2 \exp(-TI/T_1))$
  - Two unknown parameters (S_0, the signal following full recovery, and T_1) are estimated in the fitting.
  - To allow for imperfect inversion a third unknown parameter, λ, can be introduced:
  $S = S_0 (1 - λ \exp(-TI/T_1))$
  λ = 2 for perfect inversion

- For the Look-Locker approach the following three parameter model is used:
  - $S = A - B \exp(-TI/T_1^*))$
  - T_1^* is the observed recovery time and is shorter than the true T_1 due to the effect of the readout pulses.
  - T_1 is estimated by the following equation: $T_1 = (B/A-1) \, T_1^*$

Modified Look-Locker For Cardiac MR

- Multiple images are acquired at the same cardiac phase in subsequent R-R intervals after an inversion pulse. This allows images with differing inversion time (TI) values to be acquired at the same cardiac phase.

- The process is repeated multiple times with different TI values for the images acquired in the same R-R interval as the inversion pulse.

Polarity Restoration

- Signal from short TI values will originate from negative longitudinal magnetisation. However the acquired images are typically magnitude reconstructed so an algorithm is incorporated into the mapping software to determine which values should be negative.

Single Inversion Modified Look-Locker

- Following an inversion pulse one image is acquired each R-R interval to sample the longitudinal magnetisation recovery.

- The increment between consecutive TI values is thus equal to the R-R interval.

- This provides sufficient data to estimate long (e.g. native) T_1 values. However for short (e.g. contrast enhanced) tissues M_z is mostly recovered by the second RR interval and so T_1 estimation will have low accuracy as the initial part of the recovery is not adequately sampled.

Modified Look-Locker Inversion Recovery
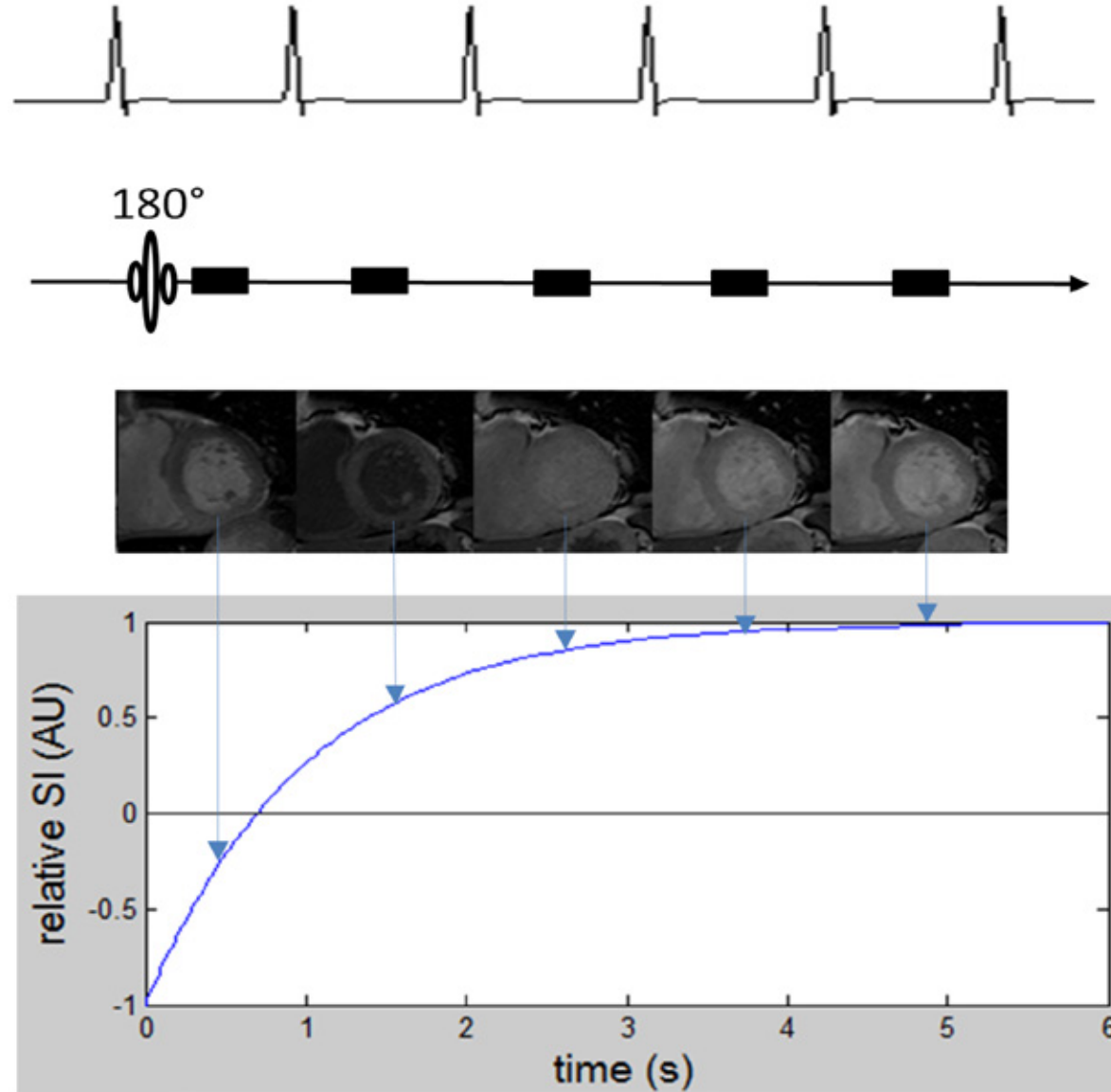
Multiple Inversions

- By performing consecutive modified Look-Locker acquisitions (with different TI values for the first image after each inversion pulse) a greater density of data can be acquired in the early part of the longitudinal magnetisation recovery. Time must however be allowed between inversions so that the magnetisation can recover to close to its equilibrium state before the next inversion.

Original MOLLI Scheme

- The originally proposed Modified Look-Locker Inversion Recovery (MOLLI) scheme acquired 11 images during 17 RR intervals with 3 inversion pulses
    - Inversion 1 – 3 images followed by 3 recovery RR intervals
    - Inversion 2 – 3 images followed by 3 recovery RR intervals
    - Inversion 3 – 5 images

- This scheme can be described using the following notation:
    - 3(3)3(3)5
    - Where numbers in brackets indicate recovery R-R intervals where no data is acquired.

3(3)3(3)5 Scheme
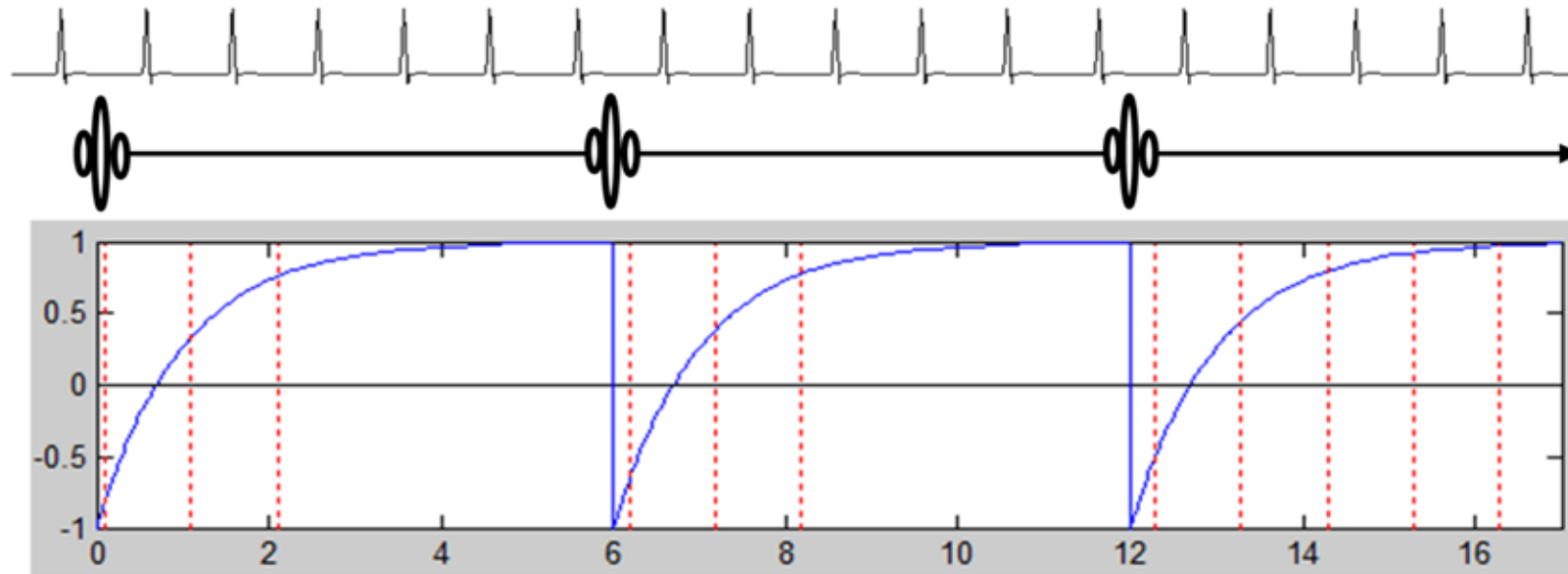
Merged Data
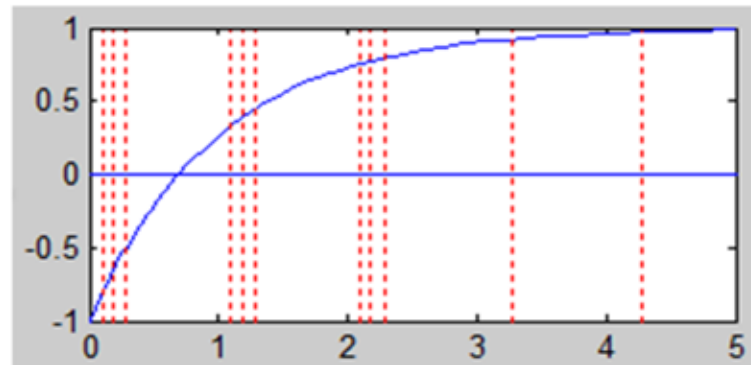
Original Scheme Limitations

- Conventional MOLLI has several limitations including:
  - Breath-hold duration – 17 beats may not be tolerable in some patient groups and so breathing motion is likely.
  - Underestimation of long T_1 – the original underestimates long T_1 values.
  - Heart rate sensitivity – the accuracy of the method varies with heart rate, underestimation is more severe at higher HR.

Shorter Schemes

- Alternative schemes with shorter breath-holds have been proposed. The optimal scheme choice depends on the expected T_1 values. For long (e.g. native) T_1 it is necessary to leave sufficient time between inversion pulses for near complete longitudinal recovery. For short (e.g. contrast enhanced) T_1 longitudinal recovery is achieved more quickly but a greater density of data at short TI values (within the same R-R interval as inversion) is required. Various shortened schemes have been proposed including the following 11 beat schemes:
  - 5(3)3 – for native T_1
    - 8 beats between inversions, 2 samples with short TI
  - 4(1)3(1)2 – for contrast enhanced T_1
    - 4 or 5 beats between inversions, 3 samples with short TI

ShMOLLI

- A 9 beat scheme (5(1)1(1)1), dubbed Shortened MOLLI or ShMOLLI, that can characterise short and long T_1 values has also been proposed. Use of data from the second and third inversions would lead to errors in estimation of long T_1 values due to incomplete longitudinal recovery. A conditional fitting algorithm, which only uses all data for short T_1 values, is therefore required.

ShMOLLI Conditional Fitting

T_1	Inversions To Use	Reason
Long	First only (5 images)	Longitudinal magnetisation will not have recovered by the second inversion
Short	First two (6 images)	Longitudinal magnetisation will have recovered by the second inversion but will not recover between the second and third
Very short	All (7 images)	Longitudinal magnetisation will have recovered by the second and third inversions

Background

- T_1 reduction in myocardium after contrast agent administration depends on the concentration of contrast agent present in the extracellular volume.

- The relative difference in extracellular volume fractions of blood and myocardium can be estimated by quantifying the relative changes in T_1.

Method

- T_1 is measured before and after contrast agent administration in blood and myocardium by T_1 mapping.

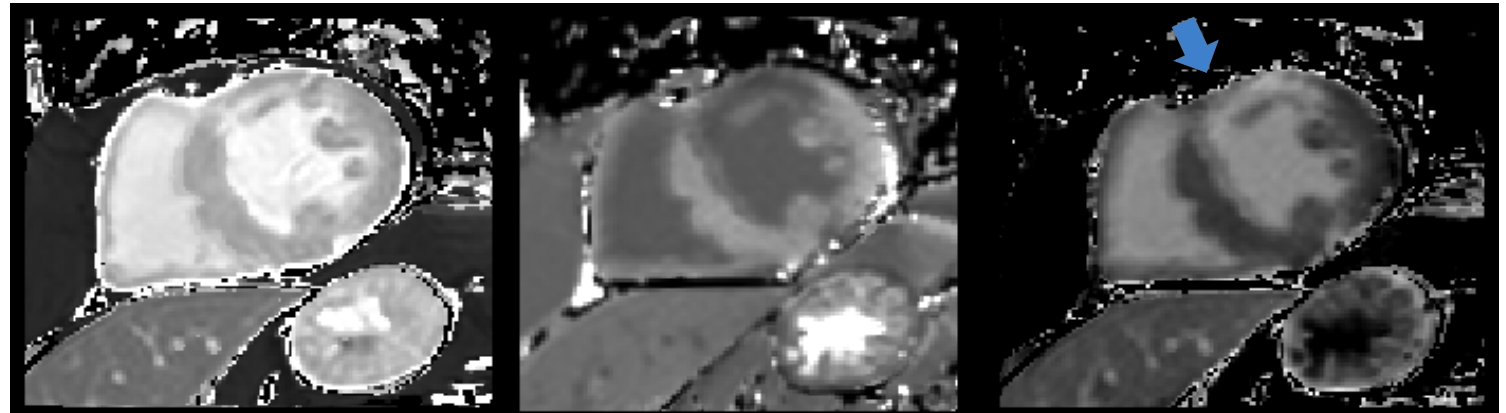- The partition coefficient, λ, is calculated using the following formula:

$$λ = (Change\ in\ myocardial\ R_1)/(Change\ in\ blood\ R_1)$$

- ECV is calculated from λ and hematocrit (Hct) by the following formula:

$$ECV = (1-Hct)\ λ$$

Left to right: Native T_1 map, Contrast enhanced T_1 map and calculated ECV map. A region of elevated ECV is apparent (arrow).

Achieving Contrast Equilibrium

- The quantitative ECV measurement method is only valid if contrast equilibrium between blood and the interstitium is attained. This can be achieved by contrast infusion or, more practically, can be approximated by allowing a sufficient delay after a bolus administration. For healthy myocardium approximately 15 minutes should be sufficient, but for some disease states a longer delay may potentially be needed.

Motion

- Motion within a breath-hold may corrupt T_1 maps and inconsistent breath-hold position between native and enhanced T_1 maps may corrupt ECV maps if not corrected for. Instead of calculating ECV for each voxel it can alternatively be calculated on a regional basis, by contouring the same regions on the native and enhanced T_1 maps.

Glossary of Terms and Abbreviations

ADC	Analogue-to-digital convertor	GBCA	Gadolinium-based contrast agents	T_1	Longitudinal (spin-lattice) recovery time
B_0	External magnetic field	GE	Gradient echo	T_2	Transverse (spin-spin) decay time
B_1	Radiofrequency field	IR	Inversion recovery	T_2^*	Transverse decay including B_0 inhomogeneity effects
bSSFP	Balanced steady-state free precession	k_0	Central line of k-space		
CHESS	Chemical shift selective fat suppression	MOLLI	Modified Look-Locker inversion	TD	Trigger delay
		MR	Magnetic resonance	TE	Echo time
CSPAMM	Complementary spatial modulation of magnetisation	M_{xy}	Component of the magnetization in the xy plane	TF	Turbo factor
				TI	Inversion time
ECG	Electrocardiogram	M_z	Component of the magnetization in the z direction	TR	Repetition time
ECV	Extracellular volume			TSE	Turbo spin echo
EPI	Echo planar imaging	N_{PE}	Number of phase-encoding steps	α	Flip angle
ETL	Echo train length	PD	Proton density	γ	Gyromagnetic ratio
FE	Frequency encoding	PE	Phase encoding	VENC	Velocity encoding sensitivity
FID	Free induction decay	PPD	Preparation pulse delay	ω_0	Resonant frequency
FOV	Field of view	R_1	Relaxation rate, $1/T_1$		
FSE	Fast spin echo	rBW	Receiver bandwidth		
G_{SS}, G_{PE}, G_{FE}	Magnetic field gradient in the slice select, phase encoding and frequency encoding direction.	RF	Radiofrequency		
		ShMOLLI	Shortened MOLLI		
		t	Time		

This pocket guide is intended as a quick reference guide. For further detail we refer the reader to the following open access review articles and the articles cited within:

- **Ridgway, JP.** 2010. Cardiovascular Magnetic Resonance Physics for Clinicians: Part I. *Journal of Cardiovascular Magnetic Resonance* **12**:71.

- **Biglands, JD, Radjenovic A and Ridgway, JP**. 2012. Cardiovascular Magnetic Resonance Physics for Clinicians: Part II. *Journal of Cardiovascular Magnetic Resonance* **14**:66.

If you have any questions, suggestions or feedback, please contact:

cmr.pocket.guide@gmail.com

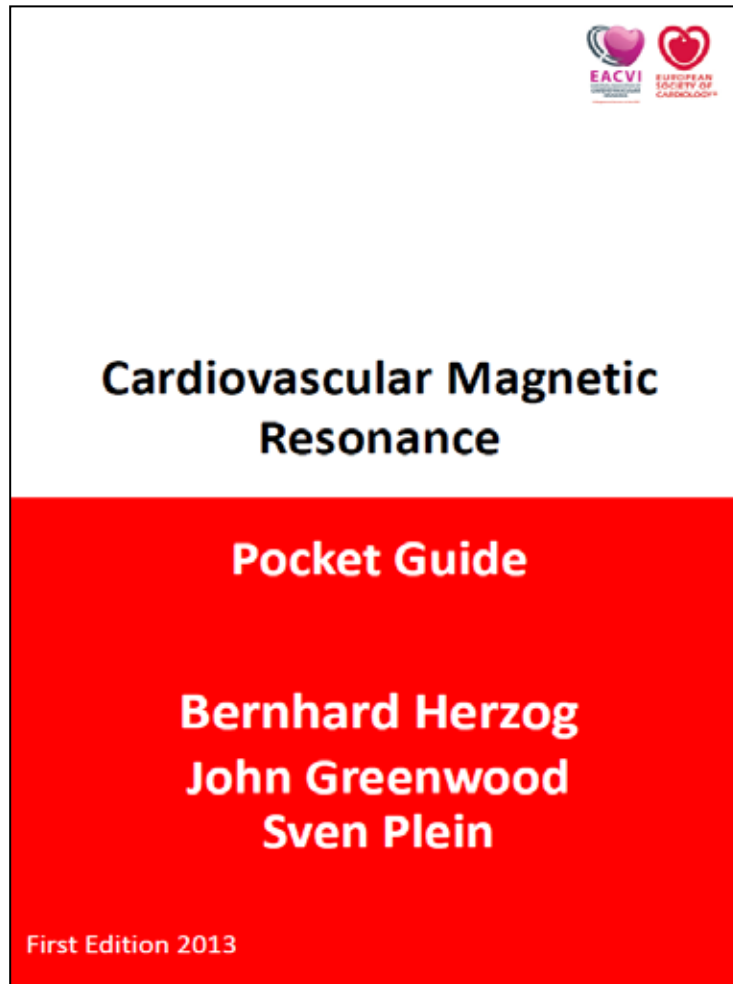| Authors: | David Broadbent [1,2] |
	Ananth Kidambi [1]
	John Biglands [1,2]
Series Editors:	Bernhard A. Herzog [1,3]
	John P. Greenwood [1]
	Sven Plein [1]

1. Multidisciplinary Cardiovascular Research Centre &
Leeds Institute of Cardiovascular and Metabolic Medicine,
University of Leeds,
Leeds General Infirmary,
Great George Street, Leeds,
LS1 3EX, United Kingdom.

2. Department of Medical Physics and Engineering,
Leeds Teaching Hospitals,
Great George Street,
Leeds,
LS1 3EX, United Kingdom.

3. University Hospital Zurich
Cardiac Imaging, C NUK 40
Rämistrasse 100
CH-8901 Zürich
Switzerland.

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