Optimizing and Manipulating Blood-Tissue Contrast in Cardiac MRI

Zhongwei Zhang, MD, PhD; Jie Zheng, PhD; Scott M. Bugenhagen, MD, PhD;

Manish Aggarwal, MD; Pamela K. Woodard, MD

Mallinckrodt Institute of Radiology, Washington University School of Medicine, St Louis, USA

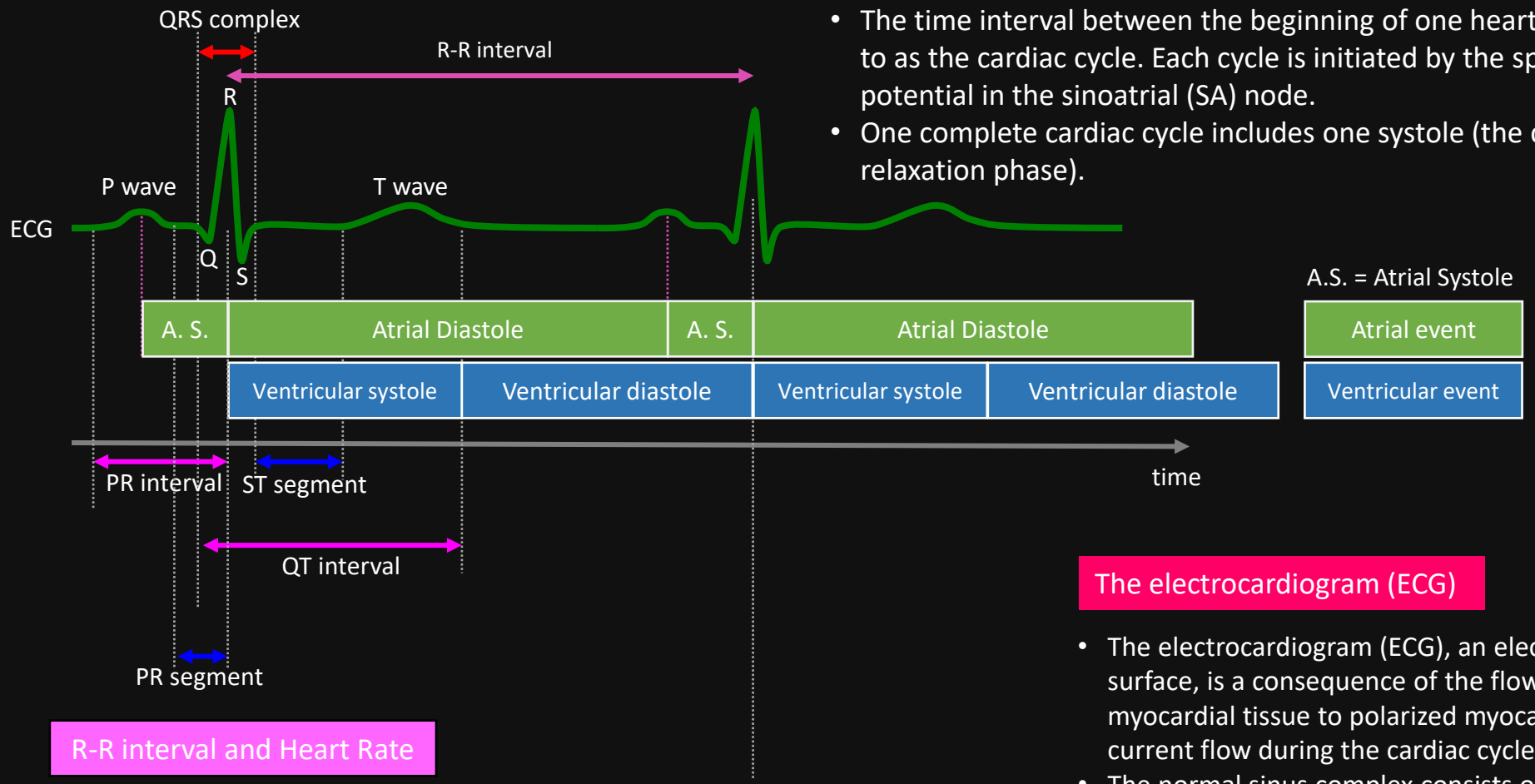
Introduction

- Cardiac MRI (CMR) excels in distinguishing blood from surrounding tissues, providing critical insights into a wide range of cardiac and vascular conditions. This ability stems from both intrinsic MR properties and advanced acquisition strategies that enhance or suppress blood signal to meet specific clinical objectives.

- This Education Exhibit will present a comprehensive overview of blood–tissue contrast mechanisms in cardiac MRI, focusing on the principles, technical foundations, and clinical applications of key imaging techniques:
  - Cardiac Cycle, Synchronization, and MRI Data Acquisition – Fundamental concepts of cardiac gating and synchronization to optimize temporal resolution.
  - Motion Correction and Navigation – Techniques to mitigate motion artifacts and improve image quality.
  - Blood and Tissue MR Properties and Contrast Mechanisms – Review of intrinsic properties of blood and myocardium, including flow effects, and their influence on MRI contrast.
  - Black-Blood, Gray-Blood, Bright-Blood, and Simultaneous Black and Bright Blood Imaging – Principles and applications of various blood suppression and enhancement strategies in CMR.
  - Data Acquisition Strategies – Overview of Cartesian and non-Cartesian sampling methods (radial, spiral, hybrid) and physiological controls (ECG, iNAV, PACE, multitasking) to optimize blood contrast.
  - MRI Acceleration Techniques – Discussion of GRAPPA, CAIPIRINHA, compressed sensing, and AI-driven acceleration approaches for rapid, high-quality imaging.
  - Recent Advances in CMR – Introduction to “one-stop shopping” and push-button cardiac MRI protocols, integrating advanced contrast manipulation for comprehensive clinical evaluation.

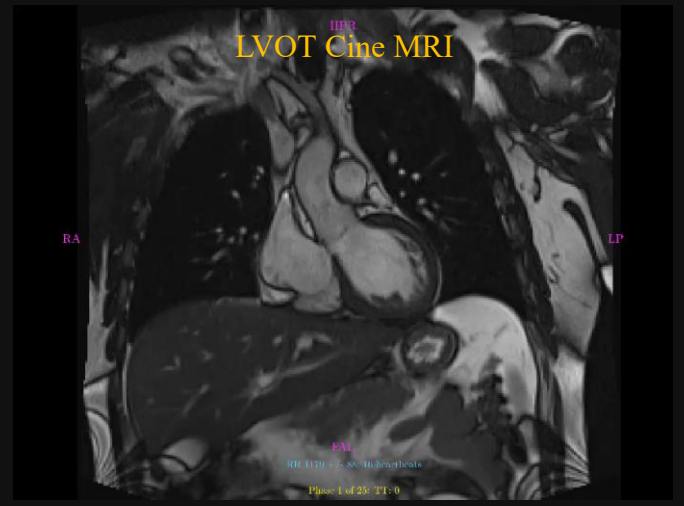
- This structured overview will equip learners with a deeper understanding of how blood–tissue contrast in CMR can be strategically optimized to improve image quality, diagnostic accuracy, and clinical efficiency.

Cardiac Cycle and the ECG



Cardiac Cycle

- The time interval between the beginning of one heartbeat and the start of the next is referred to as the cardiac cycle. Each cycle is initiated by the spontaneous generation of an action potential in the sinoatrial (SA) node.
- One complete cardiac cycle includes one systole (the contraction phase) and one diastole (the relaxation phase).



R-R interval and Heart Rate

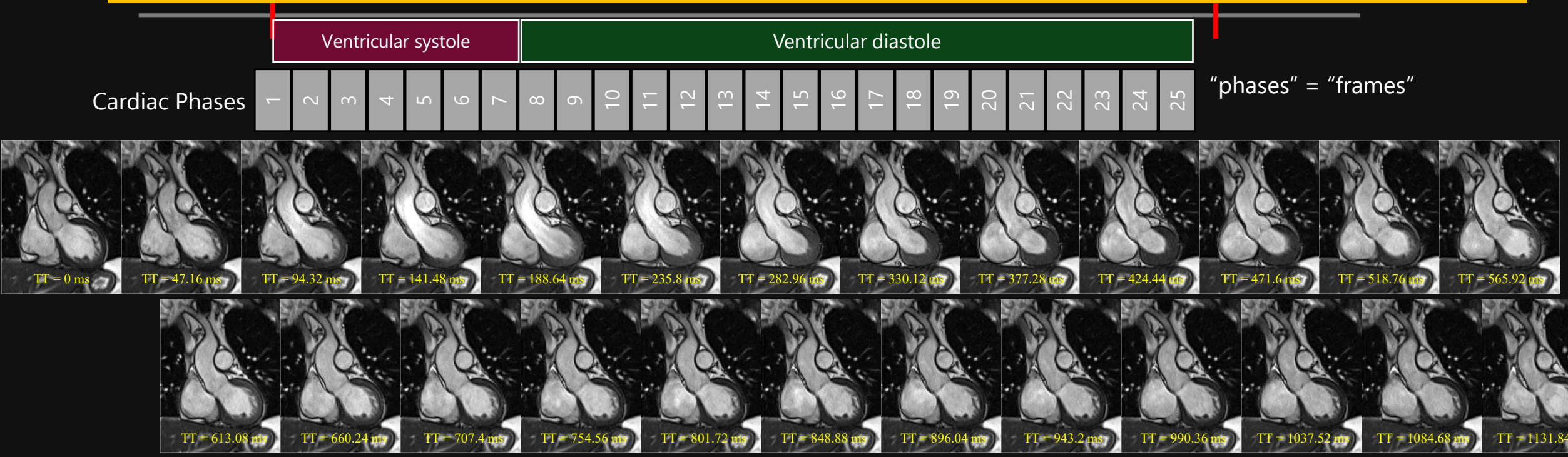
- R-R interval: the time elapsed between two successive R-waves of the QRS signal on the electrocardiogram.
- Heart rate (HR) refers to the number of times the heart beats per minute (bpm), given by: $\text{HR (bpm)} = \frac{60{,}000\ (ms/\text{min})}{\text{RR interval (ms)}}$ or $\text{RR interval (ms)} = \frac{60{,}000\ (ms/\text{min})}{\text{HR (bpm)}}$

The electrocardiogram (ECG)

- The electrocardiogram (ECG), an electrical potential recorded on the body surface, is a consequence of the flow of current from an area of depolarized myocardial tissue to polarized myocardium. The ECG provides a 1D view of the current flow during the cardiac cycle recorded against time
- The normal sinus complex consists of a P wave, a QRS complex and a T wave. What this represents in real terms is depolarization and repolarization of the heart, shown as an electrical stimulus.
- A heartbeat is represented by a series of waves that show how the heart muscle contracts and relaxes over time. The largest deflection on an ECG is often the R wave, this represents the main muscle of the heart contracting.

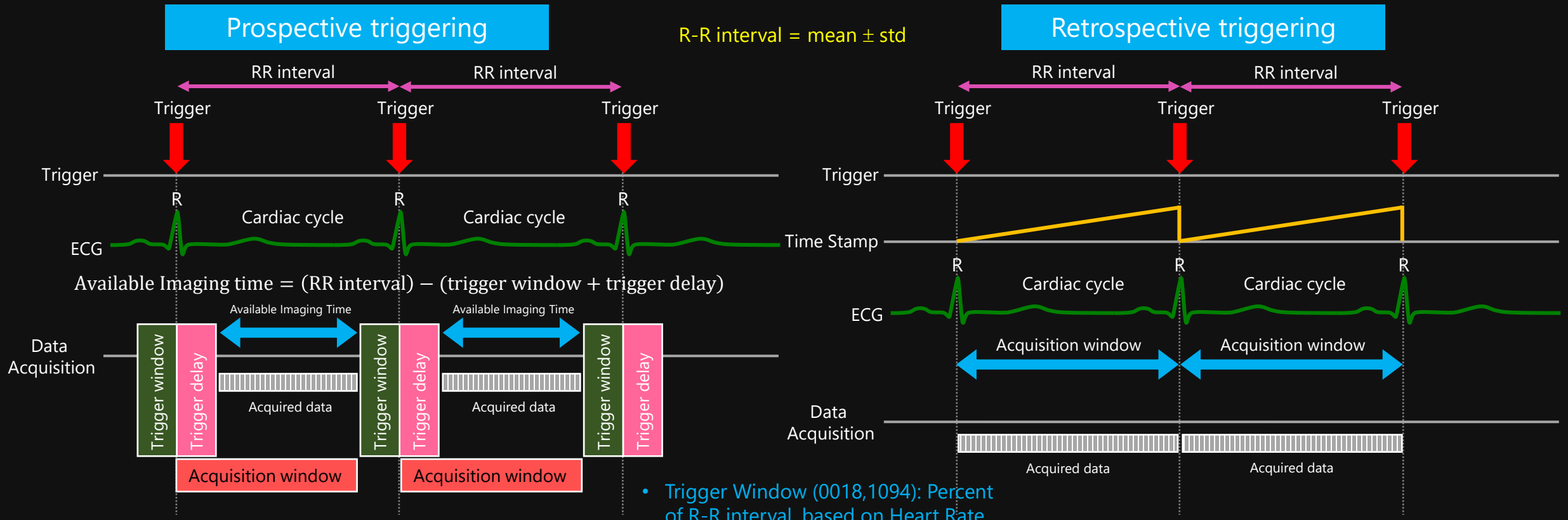
Cardiac Cycle and Phases

- In CMR, cardiac phases refer to images of the heart captured at multiple time points throughout the cardiac cycle.
  - The number of phases that can be acquired within a single RR interval is constrained by both the temporal resolution of the imaging acquisition and the duration of the RR interval (i.e., the patient's heart rate).
  - The number of acquired phases can be calculated as: #Phases = (Available RR Interval) / (Temporal Resolution)
- Temporal resolution refers to the time required to capture a single image at one time point in the cardiac cycle or the time between consecutive phases.
  - A high temporal resolution (small values) allows for a more precise and detailed representation of the heart's dynamic motion across more phases of the cardiac cycle.



LVOT Cine MRI as an example: 2D True FISP Cartesian acquisition with parallel imaging (GRAPPA 2), RR = 1179±88 ms; 14 segments, 16 heartbeats

Cardiac Synchronization: The Basics

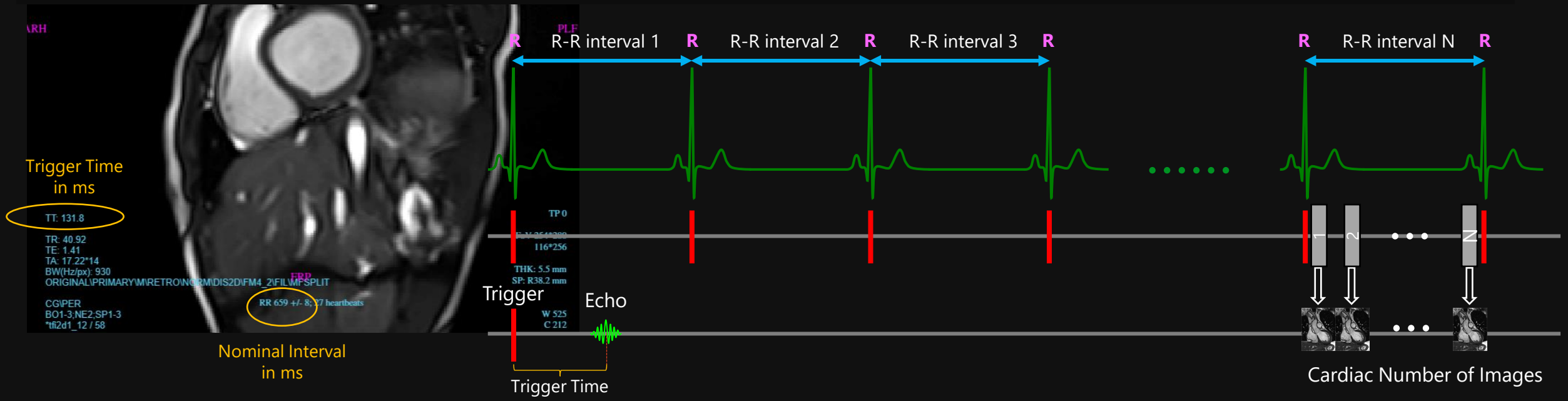


- The acquisition window is usually the average captured cycle, with a constant (e.g., 60 ms) populated.
- Trigger Window (0018,1094): Percent of R-R interval, based on Heart Rate (0018,1088), prescribed as a window for a valid/usable trigger.

- In ECG gating, timing refers to the precise synchronization of the MRI data acquisition with specific events in the cardiac cycle, based on the electrical activity of the heart as captured by the ECG signal.
- Prospective gating: Data Acquisition take place after predetermined delay from trigger signal, R wave triggers the MR pulse sequence to acquire a fixed number of data lines at a given time point within the R-R interval. Data is prospectively collected at a particular time in the cardiac cycle with respect to the R-wave detection. 1) Cover less than entire cardiac cycle; 2) Sensitive to arrhythmia; 3) Acquisition window manually adjusted;
- Retrospective gating: The timing of data acquisition is arbitrary and detected R-waves as reference points. Data is collected continuously over several heart beats and acquired data is sorted into different cardiac phase determined by the delay from trigger signal. 1) Measures through entire cardiac cycle; 2) Arrhythmia rejection is available; 3) Acquisition window automatically adjusted

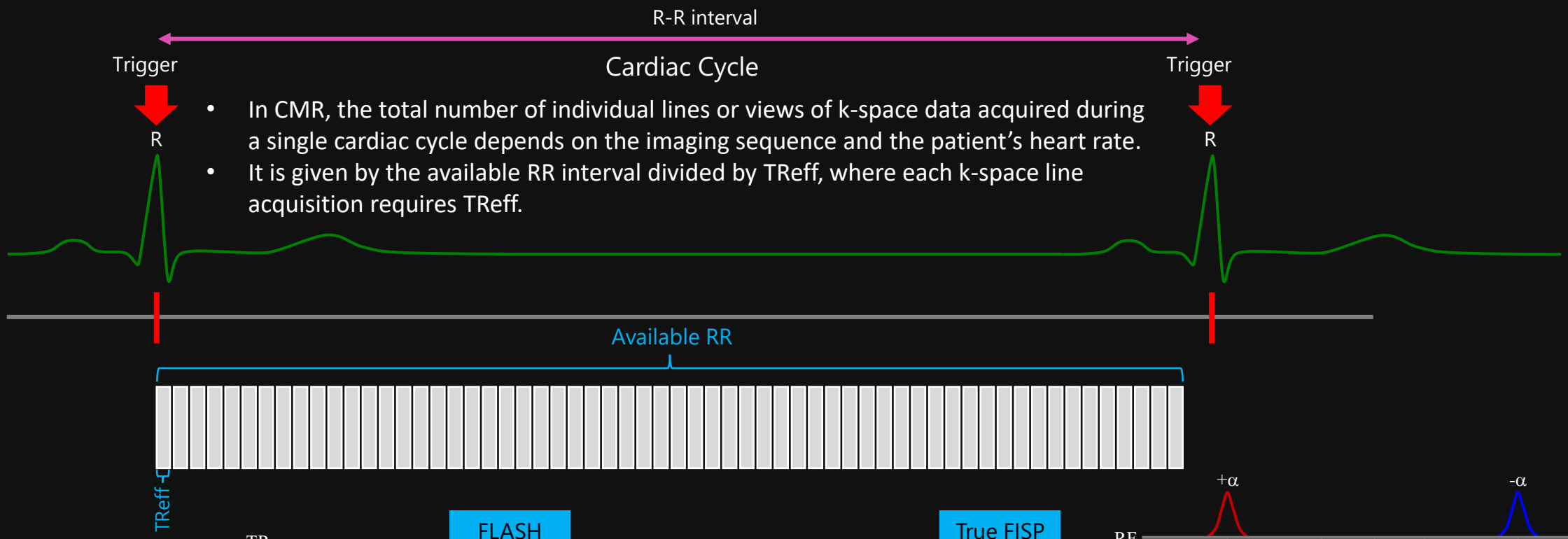
Understanding Cardiac Timing Relationships

- Cardiac MR is distinct from other imaging studies due to its close synchronization with the cardiac cycle, making precise timing essential.
- Understanding cardiac timing in both prospective and retrospective ECG gating is crucial for accurately linking images to specific cardiac phases, ensuring physiologic consistency and reliable quantitative analysis.
- Trigger Time and Trigger Delay define when data acquisition occurs relative to the ECG R-peak.
- R–R Interval tags (nominal vs. actual) define the cardiac cycle duration, which determines temporal alignment across frames.
- Trigger Window and Cardiac Number of Images control the tolerance and temporal resolution, affecting how finely the cardiac cycle is sampled.

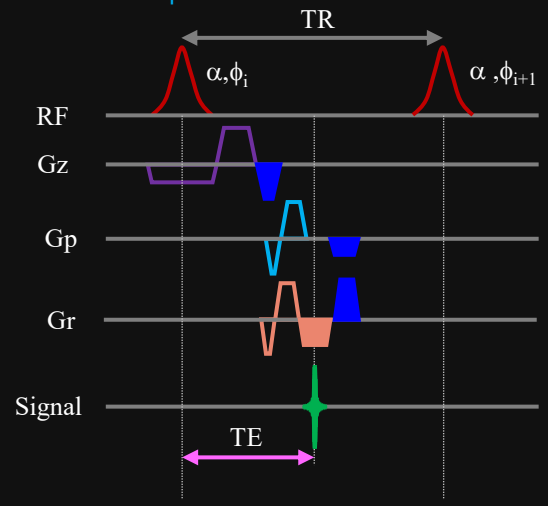


DICOM Tag	Attribute Name	Definition	Timing Source	When Measured / Applied	Purpose
(0018, 1060)	Trigger Time	Time (in ms) between the peak of the R-wave and the peak of the echo produced. In segmented k-space, TE(eff) is the time between the peak of the echo that is used to cover the center of k-space.	ECG R-wave (physiology)	Per acquisition	Defines time delay between ECG trigger and data sampling center. Other related tags include Nominal Cardiac Trigger Time Prior to R-peak (0020,9154), Actual Cardiac Trigger Time Prior to R-peak (0020,9155)
(0018, 1062)	Nominal Interval	Average R–R interval used for the scans, in ms. Average R–R interval = (Sum of all R–R intervals) / (Number of R–R intervals measured).	Average ECG R–R (planning)	During scan setup or early acquisition	Sets nominal cardiac cycle length used for sequence timing. Other related tags include Cardiac RR Interval Specified(0018,9070), R–R Interval Time Nominal (0020, 9251)
(0018, 1090)	Cardiac Number of Images	Number of images (phases/frames) per cardiac cycle.	Sequence definition	Sequence setup	Determines number of temporal frames reconstructed per heartbeat.
(0020, 9153)	Nominal Cardiac Trigger Delay Time	Nominal time in ms from the previous R-peak to the Frame Reference DateTime, expressed as a positive value.	Sequence preset (planned)	During sequence planning	Sets nominal trigger delay from R-peak to data acquisition start. Other related tags include Actual Cardiac Trigger Delay Time (0020, 9252)

Cardiac Cycle and MRI Data Acquisition: Basic Pulse Sequences



- In CMR, the total number of individual lines or views of k-space data acquired during a single cardiac cycle depends on the imaging sequence and the patient's heart rate.
- It is given by the available RR interval divided by TReff, where each k-space line acquisition requires TReff.

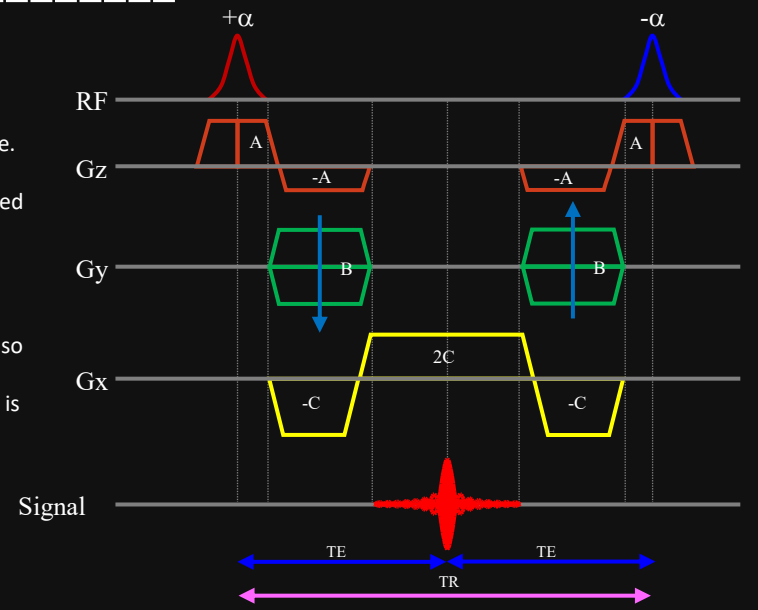


FLASH

- Spoiled gradient echo sequence with short TR, short TE and larger flip angle;
- T1 contrast;
- Compared to True FISP readout, FLASH readout is not sensitive to signal drop out from off-resonance and/or fast flow, particularly in pediatric imaging setting
- Used for 1) high-resolution display of the valves of the heart and/or flow phenomena; 2) for patients with valve replacement that could lead to strong artifacts with True FISP.

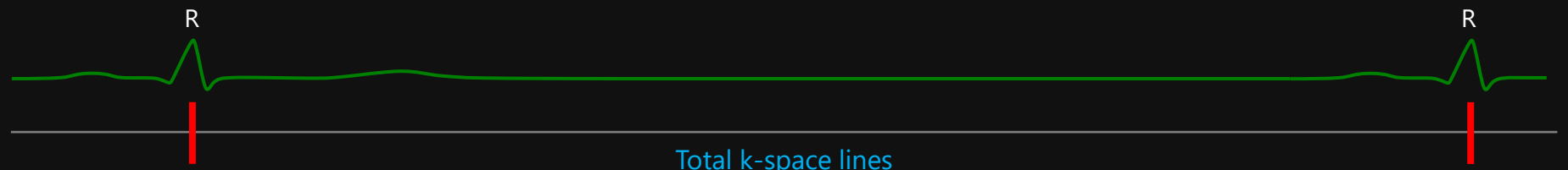
True FISP

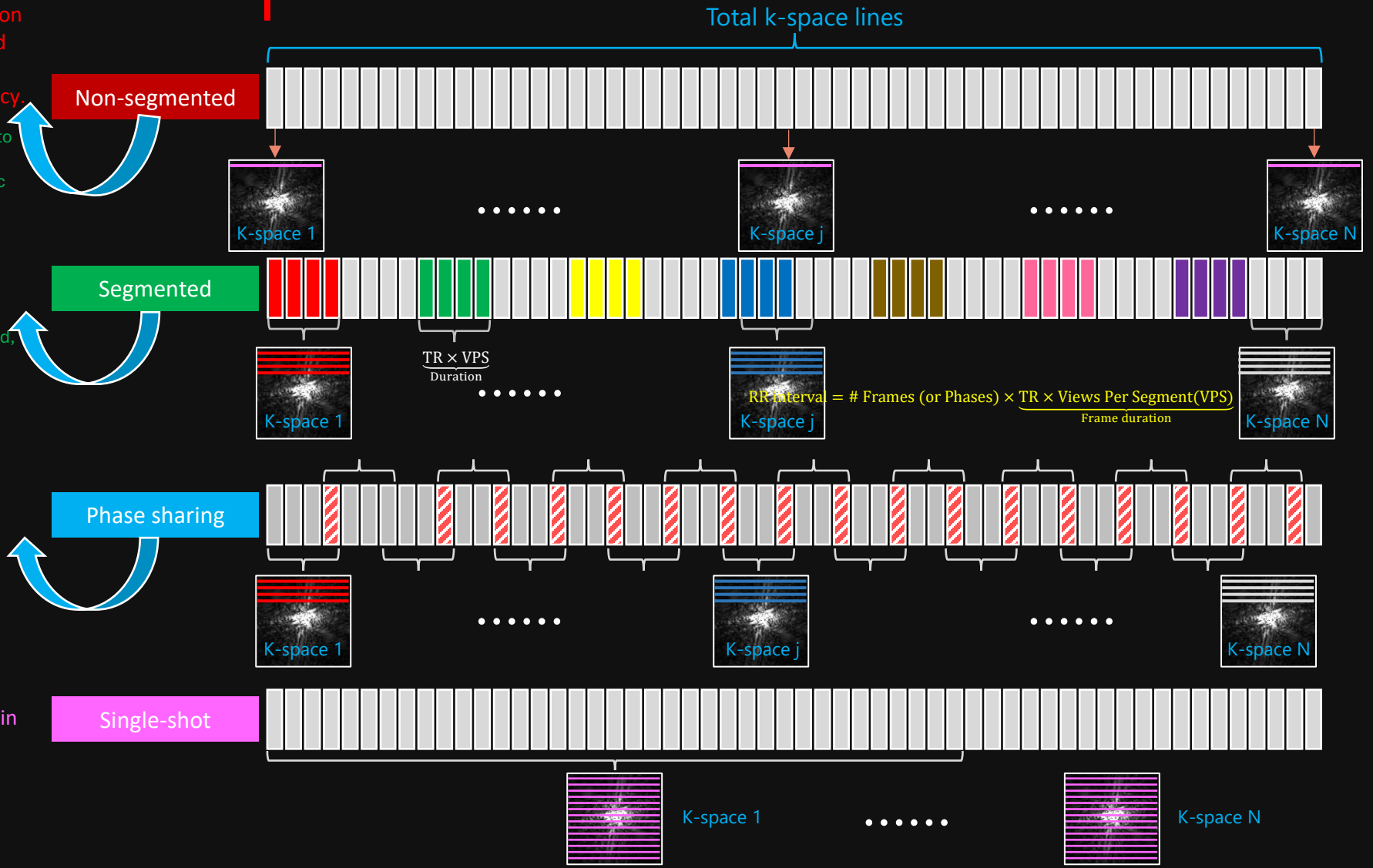
- Signal intensity depends on refocusing scheme.
- Contrast depends on T2/T1, highlighted fluid (i.e., CSF and blood) contrast: 1) Blood displayed at higher signal intensity; 2) improved blood-tissue contrast.
- Best image quality can be obtained by using shortest TR and TE, here (TR=2TE).
- Ultrafast data acquisition using very short TR, so higher temporal resolution is feasible.
- Combination with other acquisition strategies is allowable
- Used for high temporal and spatial resolution cardiac imaging



Cardiac Cycle and MRI Data Acquisition: From Raw Data to Image

- Only one k-space line is acquired per cardiac cycle.
- Need more shots per slice, the acquisition time is typically too long for breath-hold techniques.
- Rarely used in practice due to inefficiency.





- The total number of k-space lines is divided into several groups or segments.
- Multiple k-space lines are acquired per cardiac cycle for each phase. Segments are acquired repeatedly throughout the cardiac cycle, with each segment at a different temporal phase, starting from the R wave trigger. This segment acquisition continues until the next cardiac R wave trigger is encountered, at which time the next set of k-space encoding views are updated, and the process is repeated until all k-space encoding views have been acquired.
- The number of segments limits temporal resolution; more segments result in lower temporal resolution.

- Each phase acquires several previously measured k-space lines from adjacent phases.
- Fewer k-space lines are acquired per cardiac cycle.
- Acquisition can be completed within a breath-hold.

- The entire k-space filling occurs within a single cardiac cycle.

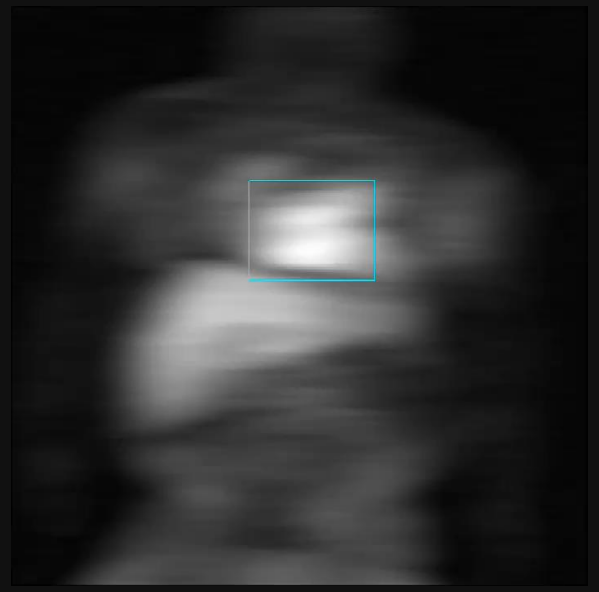
Motion in CMRI and Solutions

- Cardiac cycle motion
    - Cyclic & predictable heart motion
    - QRS complex = start of ventricular systole
    - ECG gating synchronizes data acquisition
        - K-space segments collected at identical cardiac phases across multiple heartbeats
        - Enables motion-free imaging at specific time points
- Respiratory cycle motion
    - Breath-hold Technique
        - Short acquisition window (10–20 sec)
        - Limitations:
            - Diaphragmatic drift
            - Patient discomfort
            - Image degradation near end of hold
    - Free-Breathing Motion Compensation
        - 1D Respiratory Gating
            - Uses internal navigator-echo/external sensors/bellows
            - Acquires data only at end-expiration
            - Drawback: Reduced scan efficiency (~30-40%)
        - 2D/3D Image-Based Navigators (iNAV)
            - Principle: Low-res 2D/3D images acquired before main scan
            - Separates moving vs. static tissue
            - Advantages:
            - Motion estimation in multiple directions
            - No extra planning (same FOV/orientation as main scan)
            - ~100% scan efficiency
            - Predictable & faster scan times

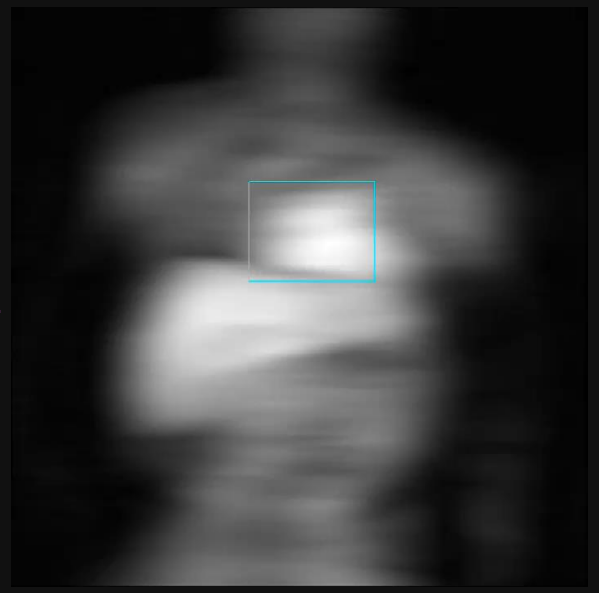
Image-based Navigator (iNAV)

- iNAV Implementation
  - Low-resolution 2D image acquired every heartbeat
  - TrueFISP: Spatial encoding of ramp-up pulses
  - Spoiled GRE: Separate low flip angle acquisition before imaging lines
- Motion Tracking & Correction
  - Beat-to-beat translation estimated in real-time
  - Primary directions: Superior-Inferior (SI) & Left-Right (LR)
  - Inline Display: Real-time motion visualization during acquisition
  - Retrospective correction: Linear phase shifts applied in k-space
- Data Normalization
  - Motion between 1st and 2nd heartbeat datasets normalized via mean motion values
  - Fixed-duration scan: Completed in predetermined number of heartbeats
- Outlier Rejection & Efficiency
  - Rejection criteria:
    - Respiratory motion outliers
    - Trigger time mismatches
  - Scan efficiency: Near 100% due to robust motion compensation
- Key Advantages
  - Real-time motion tracking
  - Retrospective k-space correction
  - High acquisition efficiency
  - Compatible with multiple sequence types

iNAV-1

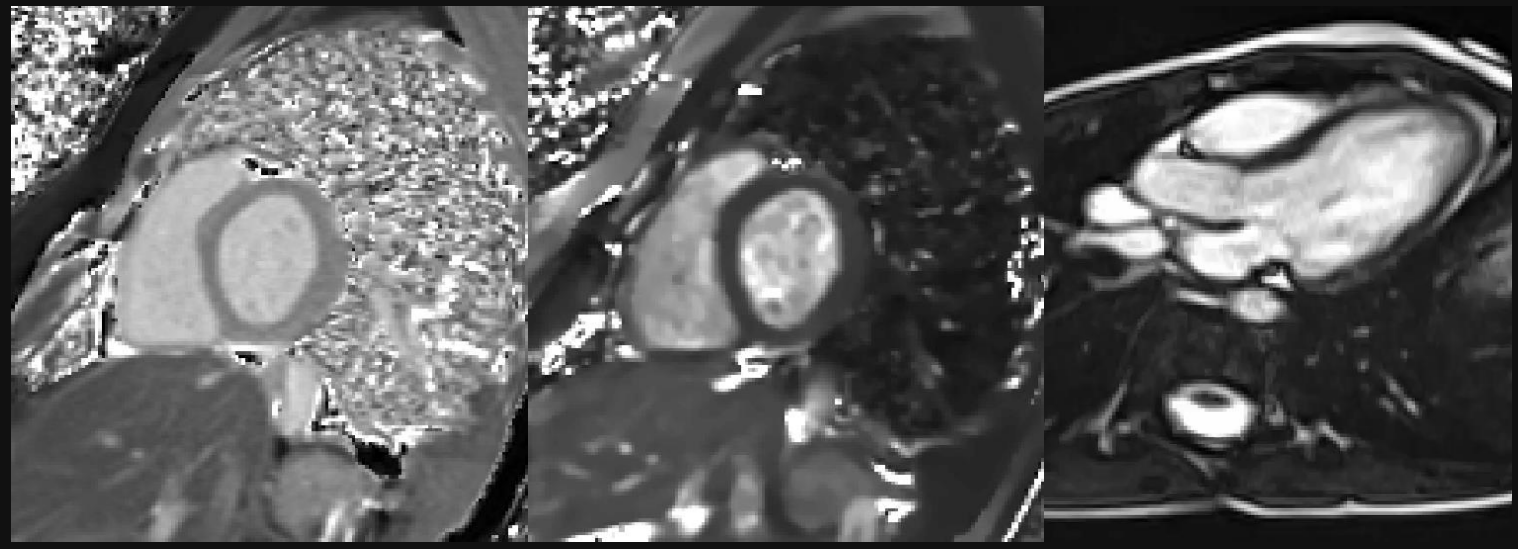


iNAV-2



Intrinsic MRI Properties of the Heart — Key Concepts

- Intrinsic tissue properties — T1, T2, T2*, proton density, T1ρ, magnetization transfer (MT), flow, susceptibility, and anisotropy — drive cardiac MRI contrast by creating distinct signal differences between myocardium and blood.
- Sequence selection (e.g., SSFP, GRE, PC-MRI) leverages these differences to enable:
  - Tissue characterization
  - Functional & hemodynamic imaging
  - Quantitative mapping (T1/T2, T2*, T1ρ, Flow)
- Contrast mechanisms
  - MT effect stronger in myocardium → improved blood–myocardium contrast
  - Flow in blood → inflow & phase contrast
  - Anisotropy → fiber orientation, DT-CMR
  - Chemical Exchange Saturation Transfer (CEST) → myocardial energetics
  - T1ρ contrast → myocardial fibrosis



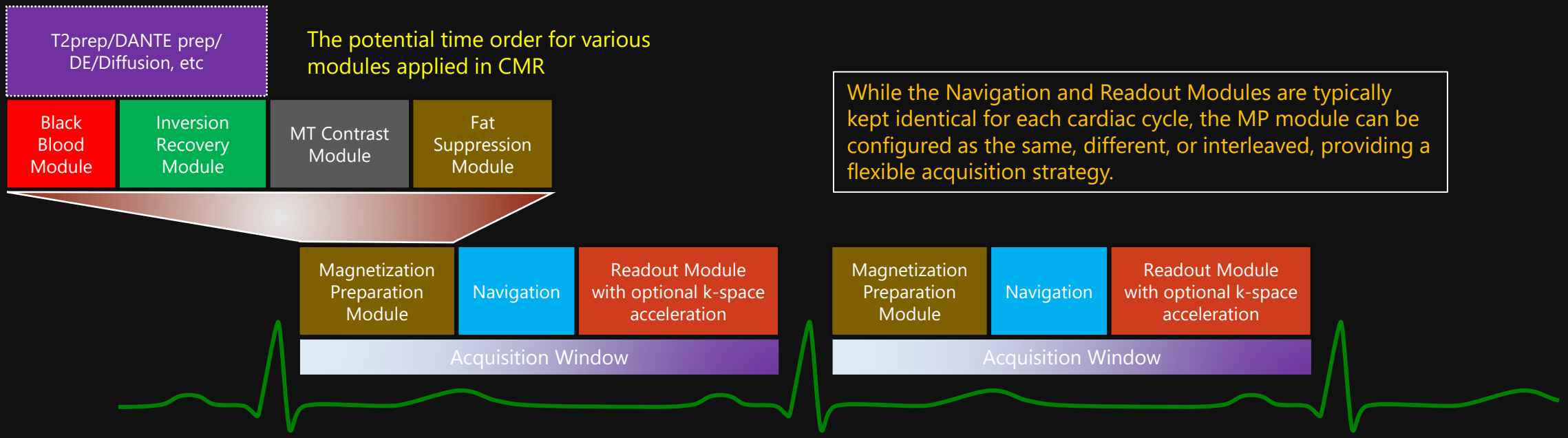
T1 map T2 map Cine

*These values can vary based on technical factors such as B0, cardiac phase, and CMR sequence, as well as patient-related factors including heart rate, age, and sex.

Property*	Myocardium (1.5 T)	Blood (1.5 T)	Myocardium (3.0 T)	Blood (3.0 T)	Clinical Use / Significance
Native T1	930-1050 ms	1200-1500 ms	1000–1100 ms	1600 – 2000 ms	Fibrosis, scar imaging, tissue characterization
Native T2	45–55 ms	200 ms	40–50 ms	180–190 ms	Edema, inflammation
T2*	25–40 ms	Variable	20–35 ms	Variable	Iron overload, susceptibility effects
MTR	High (strong MT effect) ~33%	Low (minimal MT)	High (slightly ↑ MT)	Low	Enhances blood–myocardium contrast, tissue characterization
Proton Density	Moderate	High	Moderate	High	Contributes to overall signal intensity
Flow	Contractile motion	Bulk flow	Contractile motion	Bulk flow	Phase contrast, 4D Flow MRI
Susceptibility	Mild	O_2-dependent	↑ B_0 inhomogeneity	More pronounced	BOLD MRI, oxygenation contrast
Anisotropy	Fiber orientation	Isotropic	Fiber orientation	Isotropic	DT-CMR, microstructural imaging
ECV	25–30 %	—	25–30 %	—	Myocardial fibrosis, infiltration (e.g., amyloidosis, cardiomyopathy)

Magnetization Preparation (MP) Modules in CMR

In CMR, the MP Module finely tunes tissue magnetization, reducing artifacts, improving contrast, and elevating image accuracy and diagnostic potential.



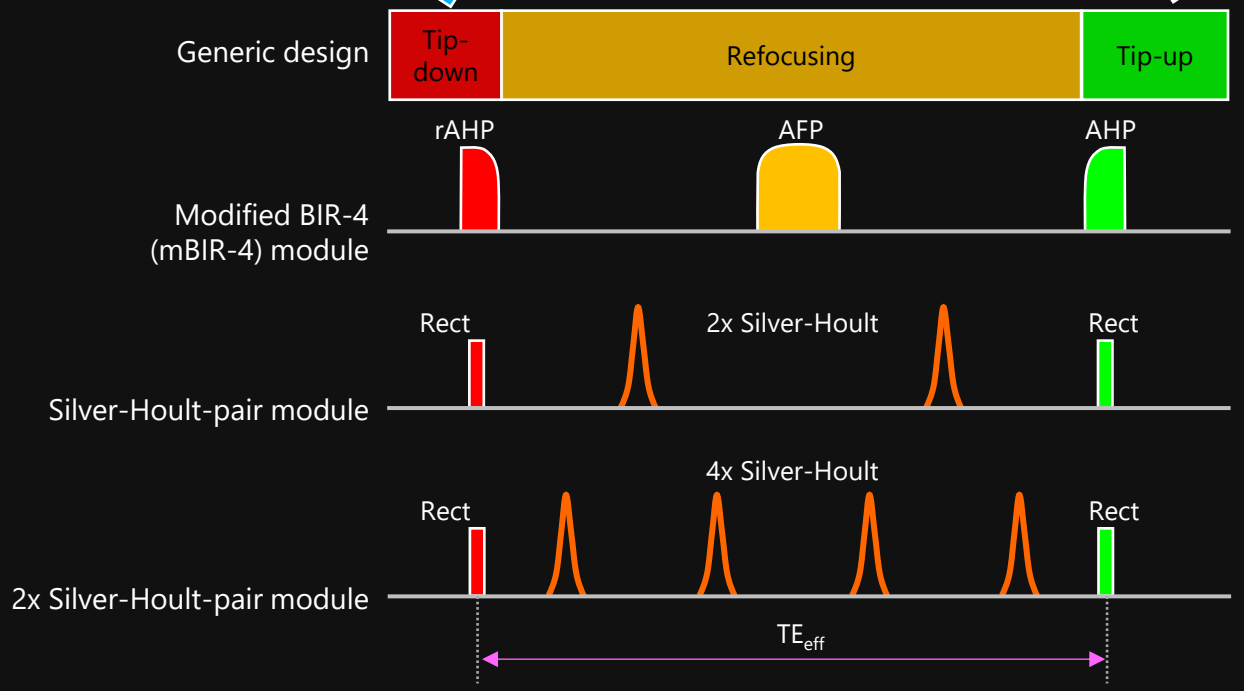
- Why are Magnetization Preparation (MP) Modules Essential in CMR?
  - Tissue contrast improvement: It enhances the contrast between different types of tissues by selectively altering magnetization, making structures and abnormalities more visible.
  - Signal Suppression: Unwanted signals or artifacts originating from specific tissues or conditions are suppressed for image quality improvement.
  - Flow and Motion Suppression: In some cases, the need to suppress signals from flowing blood is crucial to focus on the surrounding tissues.
  - MP modules can effectively reduce or eliminate these unwanted blood flow signals.

T2 preparation (T2prep) Module

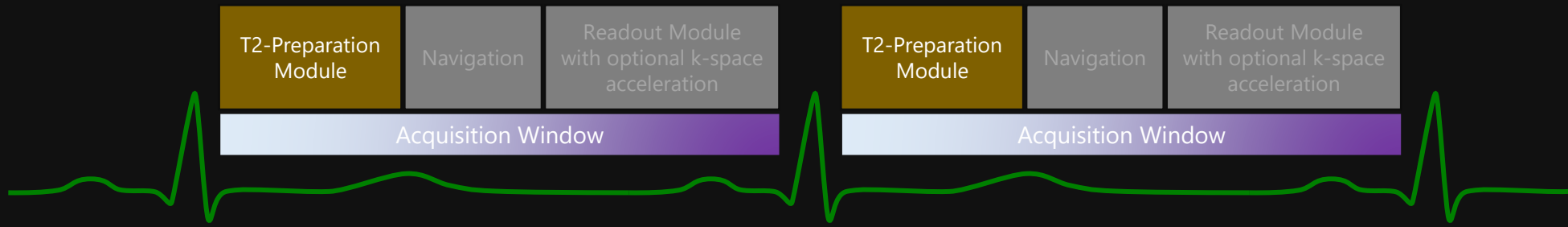
An initial tip-down (i.e., 90°x) pulse to convert M_0 into transverse magnetization

one or more refocusing pulses: a combination of delays and RF pulses designed to refocus this transverse magnetization after some signal decrease through T2 relaxation during these pulses and delays.

A final tip-up (i.e., -90° or (270°)(-360°)) pulse to return a substantial part of the refocused magnetization to longitudinal magnetization.



- A T2-preparation (T2Prep) module is a magnetization preparation technique used in MRI to create T2-weighted contrast before the main image acquisition. This method is widely applied in cardiac MRI to enhance the contrast between the myocardium and the blood pool, leveraging their intrinsic differences in T2 relaxation times.
- A standard non-selective T2Prep sequence consists of three key components: a tip-down pulse, one or more refocusing pulses, and a tip-up pulse.
- $90^{\circ}_{x} \rightarrow$ T2 decay period with refocusing pulses $\rightarrow -90^{\circ}_{x}$
  - A 90° RF pulse rotates the longitudinal magnetization (Mz) into the transverse plane (Mxy).
  - A train of 180° refocusing RF pulses compensates for static field inhomogeneities and generates a spin echo, during which transverse magnetization undergoes T2 decay.
  - A −90° RF pulse returns the residual transverse magnetization to the longitudinal axis, thereby preserving the T2-weighted contrast for the subsequent imaging readout.



Why T2prep Modules often utilize a series of refocusing pulses?

- In a conventional T2prep sequence, the goal is to generate pure T_2-weighted contrast before the main imaging readout (e.g., bSSFP, GRE). The simplest implementation would use a single 180° refocusing pulse at TE/2 to form a spin echo at TE. However, this approach is highly sensitive to magnetic field (B_0) inhomogeneities and RF transmit (B_1) nonuniformity, especially in cardiac and body MRI where the field is not perfectly homogeneous.

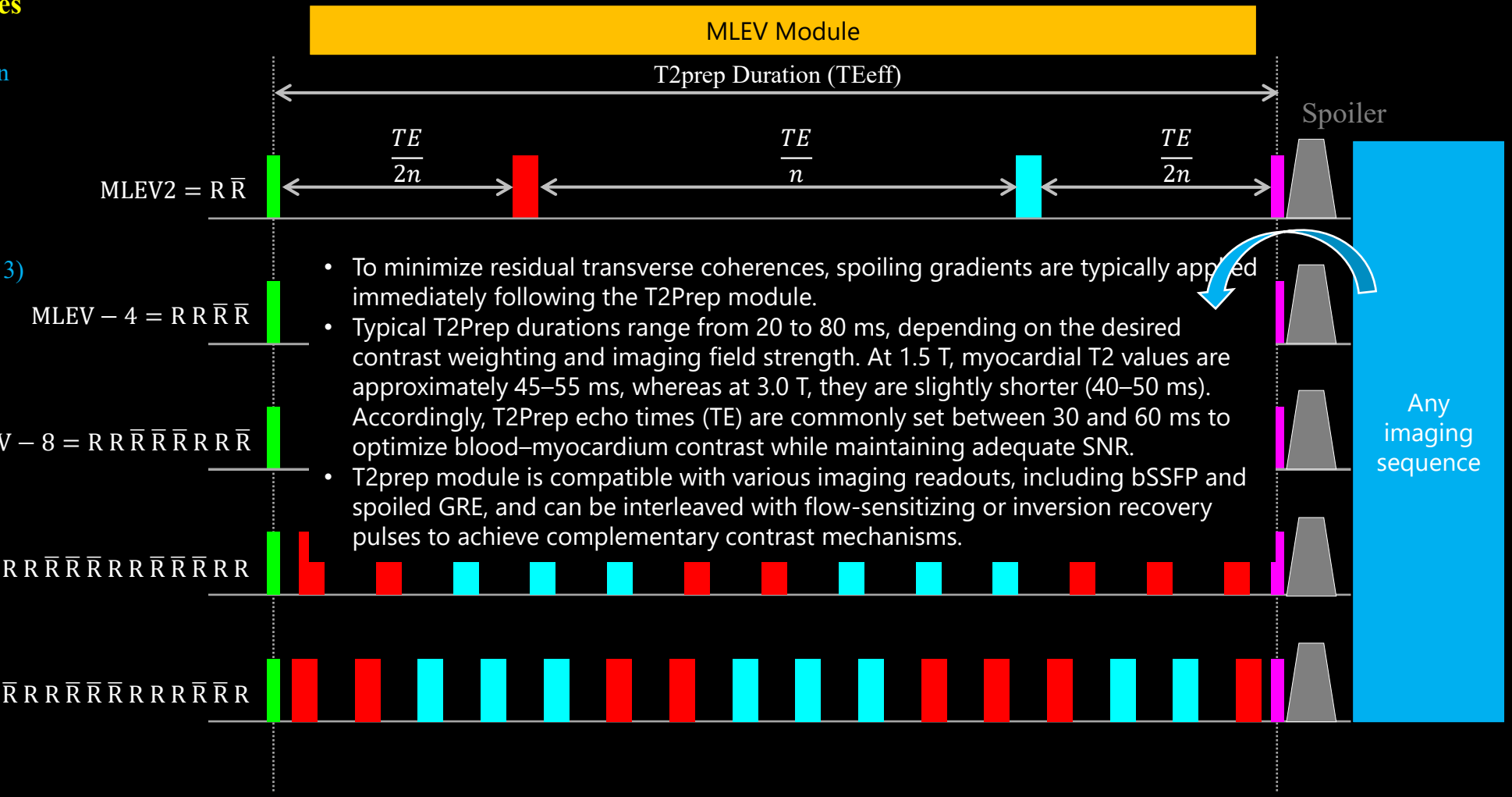
- To overcome these limitations, T2prep modules often use a train of multiple 180° refocusing pulses, such as: Carr–Purcell–Meiboom–Gill (CPMG) sequence, or Composite / adiabatic pulse trains (e.g., MLEV, BIR-4, or Hard-180 trains).

    - Compensation for B_0 Inhomogeneity
        - Small variations in the main magnetic field cause spins to precess at slightly different frequencies, leading to phase dispersion and faster signal loss (T_2* effects). Multiple 180° refocusing pulses repeatedly rephase the spins, effectively cancelling out these phase errors many times within the preparation window. This helps ensure that the observed signal decay represents true T_2 relaxation rather than additional dephasing from field inhomogeneity.
    - Reduction of B_1 Sensitivity
        - In body MRI, the transmit RF field (B_1) can vary across the imaging volume, leading to imperfect 180° flips. A single imperfect refocusing pulse (e.g., 160° instead of 180°) causes residual transverse components and incomplete refocusing. Using a phase-cycled train of refocusing pulses (e.g., MLEV-4, MLEV-8, or composite adiabatic refocusing) averages out these flip-angle errors, producing more uniform refocusing and consistent T_2 weighting across the field of view.
    - Improved Robustness and Signal Stability
        - A single-echo T2prep module may suffer from phase inconsistencies between successive repetitions, unwanted stimulated echoes, artifacts in regions with high off-resonance. By applying multiple refocusing pulses in a CPMG-like train, the magnetization trajectory remains coherent, and the effective T_2 decay follows a well-behaved exponential. This results in more stable and reproducible signal attenuation across subjects and scanner conditions.

MLEV Pattern using Non-selective Pulses as an example

Phase Cycling of Refocusing Pulses

Malcolm–Levitt (MLEV) phase-cycling scheme is the most widely used in modern T_2-preparation modules. The MLEV approach can be implemented with a variety of RF pulse types, including: 1) Single hard (rectangular) pulses; 2) Composite pulses (e.g., MLEV-weighted combinations) for slice-selective T2prep; 3) Adiabatic refocusing pulses (e.g., BIR-4, BIREF-1) for optimal performance under strong B_1 transmit field inhomogeneities.



- To minimize residual transverse coherences, spoiling gradients are typically applied immediately following the T2Prep module.
- Typical T2Prep durations range from 20 to 80 ms, depending on the desired contrast weighting and imaging field strength. At 1.5 T, myocardial T2 values are approximately 45–55 ms, whereas at 3.0 T, they are slightly shorter (40–50 ms). Accordingly, T2Prep echo times (TE) are commonly set between 30 and 60 ms to optimize blood–myocardium contrast while maintaining adequate SNR.
- T2prep module is compatible with various imaging readouts, including bSSFP and spoiled GRE, and can be interleaved with flow-sensitizing or inversion recovery pulses to achieve complementary contrast mechanisms.

Sequence name	RF phases pattern	Description
Carr-Purcell or CPMG	$(xy)^n$	Basic alternating x–y phase sequence
XY-4	$(xyxy)^n$	Compensates for off-resonance and pulse amplitude errors
XY-8	$(xyxy\ yxyx)^n$	Higher-order error compensation
XY-16	$(xyxy\ yxyx\ \bar{x}\bar{y}\bar{x}\bar{y}\ \bar{y}\bar{x}\bar{y}\bar{x})^n$	Improved robustness to both B_0 and B_1 errors
MLEV-4	$(xx\ \bar{x}\bar{x})^n$	Malcolm–Levitt composite phase cycling for B_1 insensitivity
MLEV-8	$(xx\ \bar{x}\bar{x}\ \bar{x}xx\bar{x})^n$	Enhanced averaging of flip-angle errors
MLEV-16	$(xx\ \bar{x}\bar{x}\ \bar{x}xx\bar{x}\ \bar{x}\bar{x}xx\ x\bar{x}\bar{x}x)^n$	Comprehensive compensation for higher-order imperfections

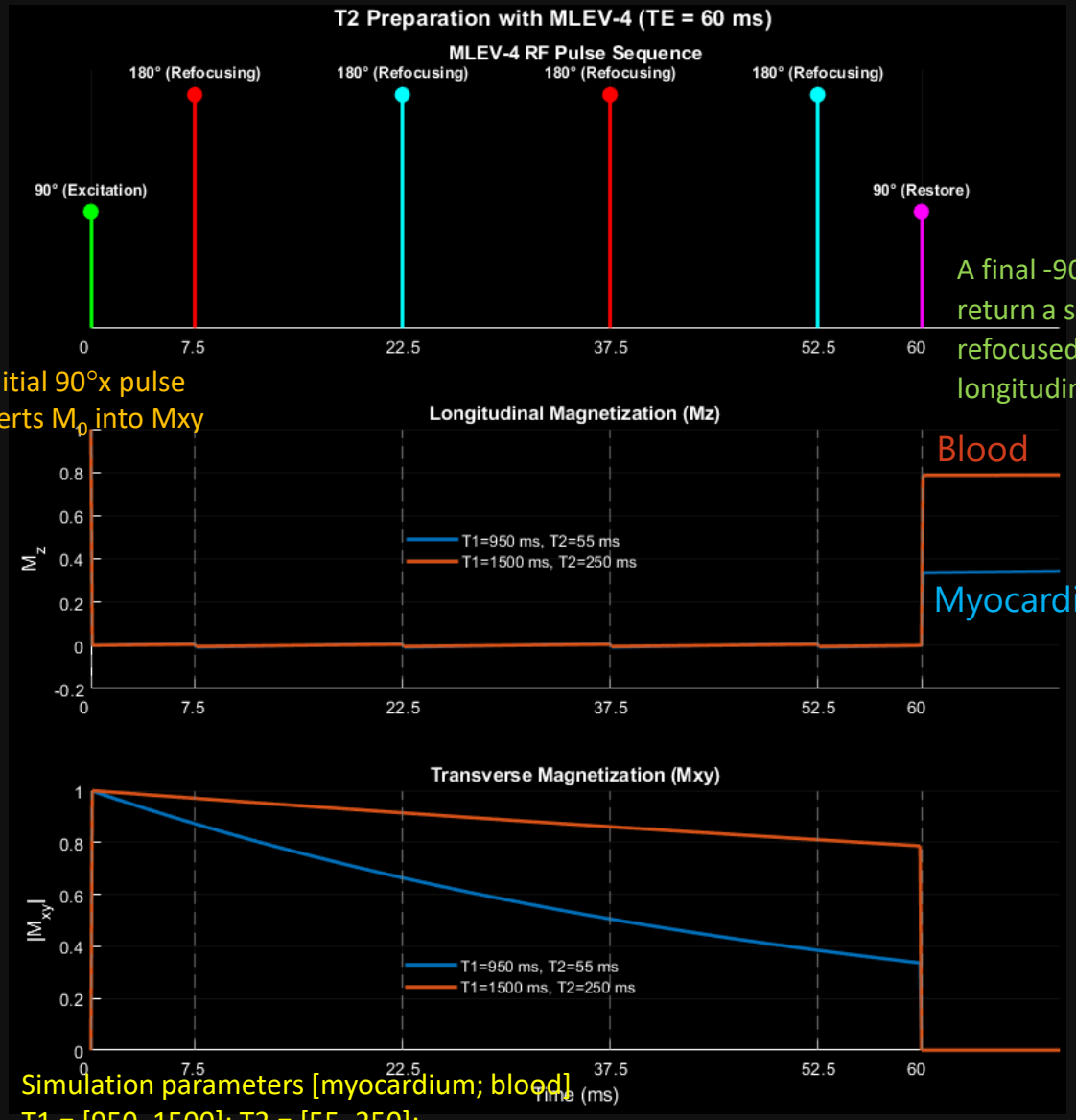
90°_x for excitation

180°_y for non-selective T2prep

-90°_x for restore

-180°_y for non-selective T2prep

T2 preparation (T2prep) Module – How does it work?

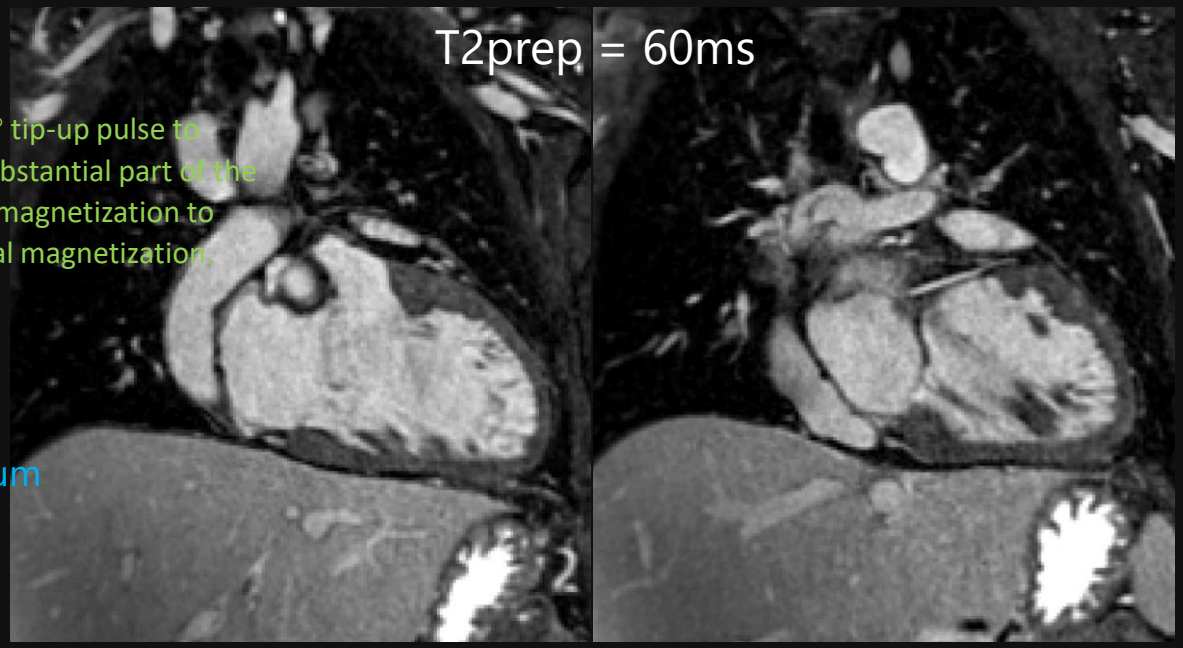


An initial 90°x pulse converts M_0 into Mxy

A final -90° tip-up pulse to return a substantial part of the refocused magnetization to longitudinal magnetization.

Blood

Myocardium



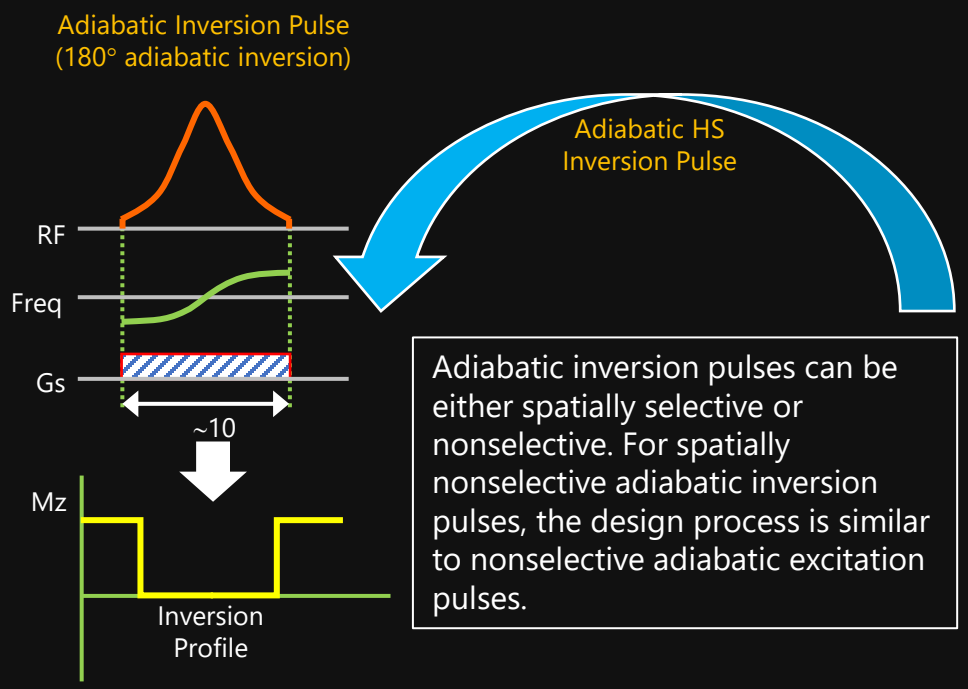
- In a T2prep module, a longer echo time (TE) leads to greater T2 weighting of the signal.
- In CMR, because the blood pool has a substantially longer T2 relaxation time than normal myocardium, its magnetization remains higher than that of myocardium immediately after T2 preparation.

Simulation parameters [myocardium; blood]
T1 = [950, 1500]; T2 = [55, 250];
TE = 60; M0 = 1; assuming spoiler gradient is applied.
For this simulation: at t = TEprep, Mz,myocardium = 0.34, Mz,blood = 0.79

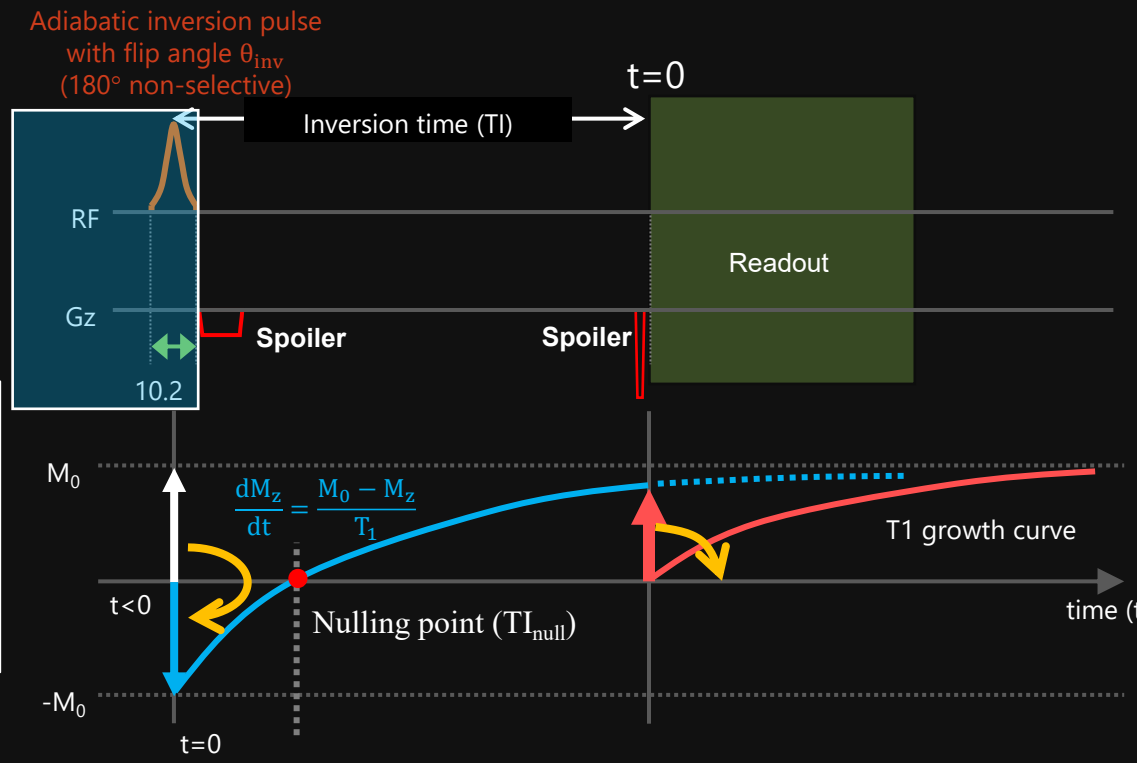
The simulation code is available at: https://github.com/sljzzw

Inversion Recovery (IR) Module

Adiabatic inversion is achieved by simultaneously modulating (i.e., varying) the amplitude and frequency of an RF pulse so that the orientation of the effective magnetic field changes from the +z to the -z axis in accordance with the adiabatic principle.



Adiabatic inversion pulses can be either spatially selective or nonselective. For spatially nonselective adiabatic inversion pulses, the design process is similar to nonselective adiabatic excitation pulses.



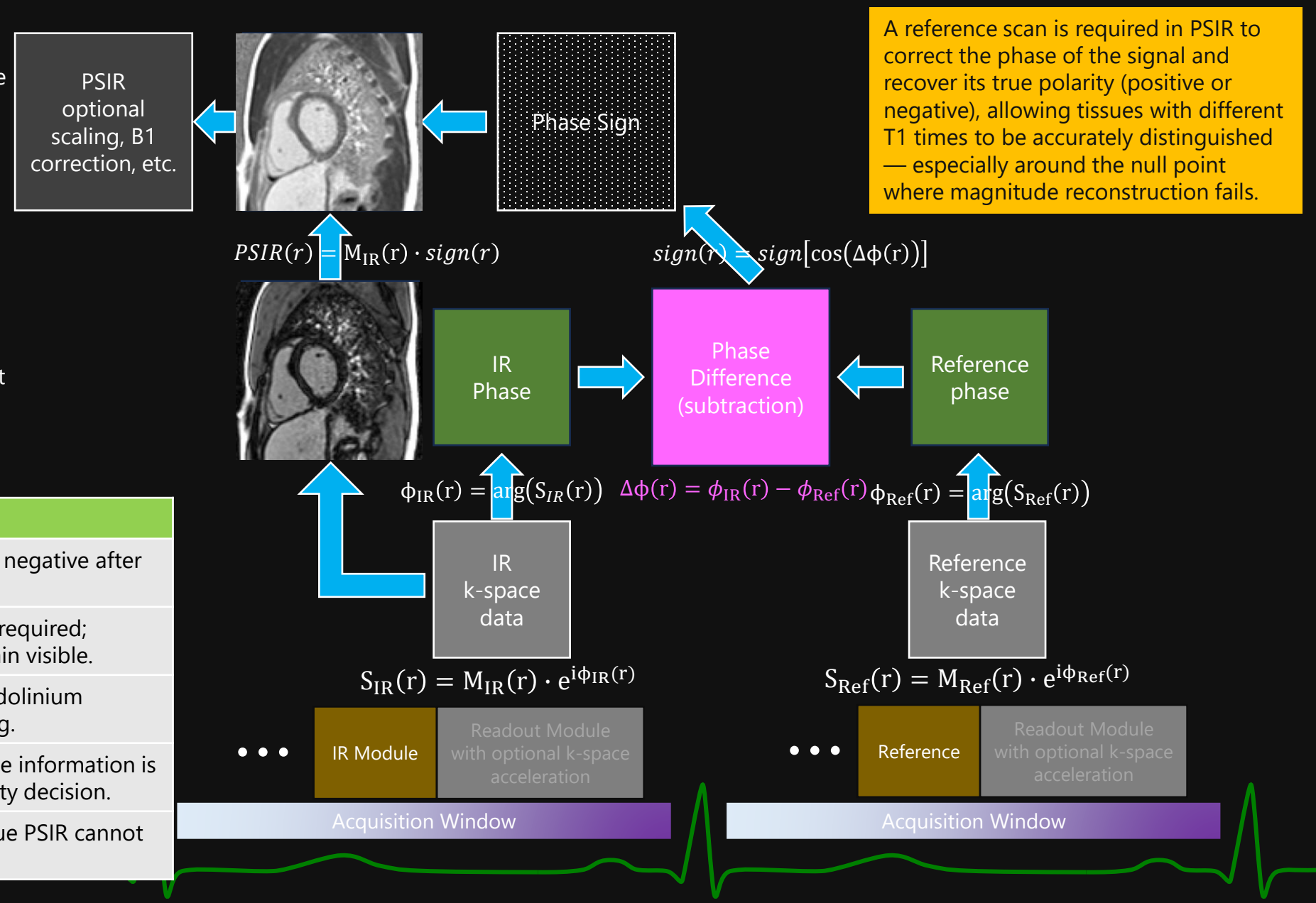
The downside of adiabatic pulses is their longer pulse duration, higher RF amplitude requirement, and increased SAR.

- Central to all the IR sequences is the application of an inversion RF pulse that flips the longitudinal magnetization from the +z to the -z direction.
- Theoretically, IR can be used to suppress any tissue of target with known T1 relaxation time by applying appropriate TI

Application	IR Variant	Purpose	Typical TI
Late Gadolinium Enhancement (LGE)	IR-GRE	Null normal myocardium to highlight infarct or scar (bright)	~250–350 ms at 1.5 T (adjusted by Look-Locker or TI-scout)
Black Blood Imaging	Double IR	Null blood signal → improve visualization of myocardium and wall	Depends on blood T1
T1 Mapping	MOLLI / ShMOLLI (modified IR)	Quantitative T1 relaxation measurement	Multiple TIs sampled
Fat suppression	STIR (Short TI IR)	Null fat signal	TI ~150–180 ms at 1.5 T
Myocardial tissue characterization	PSIR (Phase Sensitive IR)	Improves contrast robustness against TI mis-selection	Similar to LGE

Phase-Sensitive Inversion Recovery (PSIR)

- In PSIR, the magnetization can be positive or negative, depending on the TI and tissue T1 values and the time at which the readout sequence is applied. By preserving the sign of the MR signal intensity, the image contrast can be enhanced by selecting appropriate TI value.
- Tissues with more negative longitudinal magnetizations appear darker than those with less negative or more positive magnetizations. Since the data is reconstructed with the obtained phase information, the images are less dependent on an optimized inversion time.

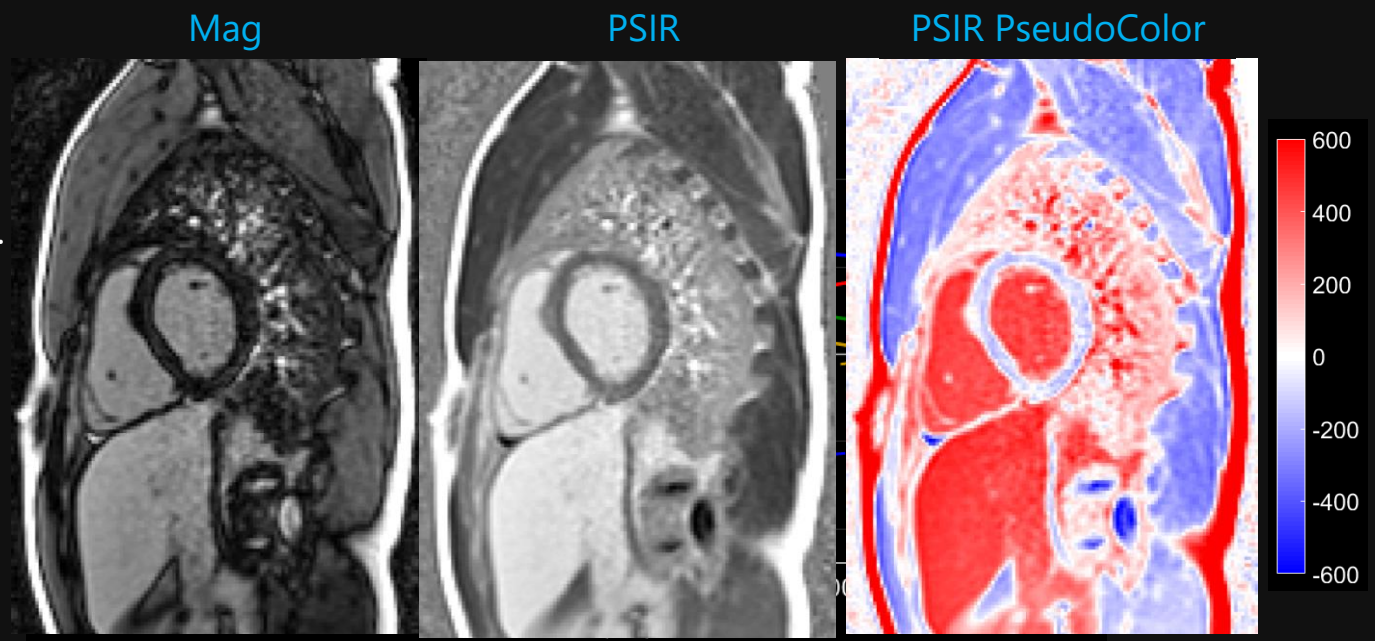


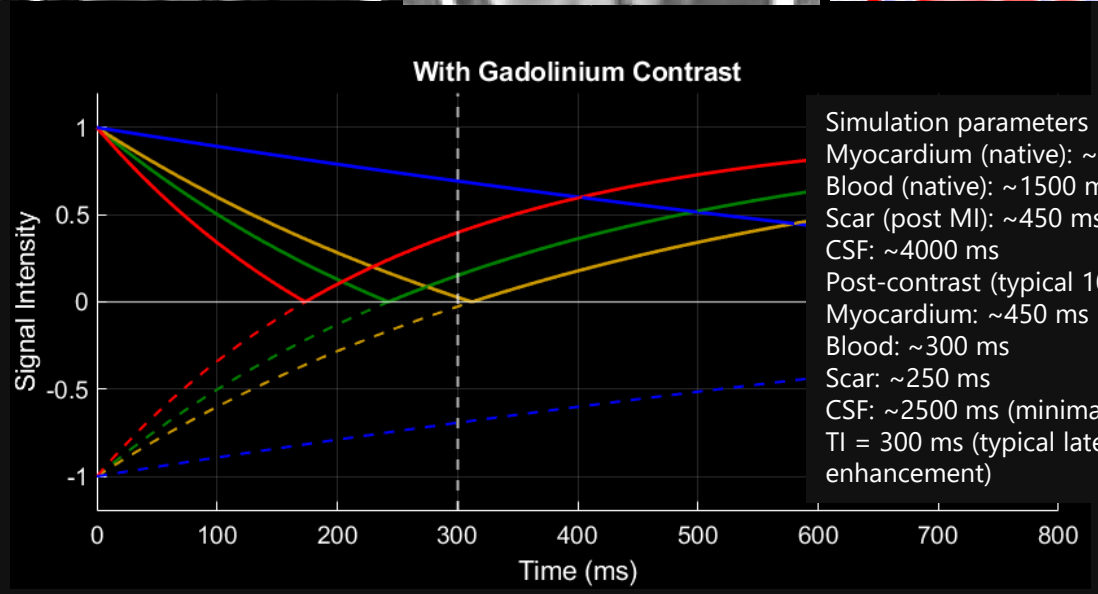
A reference scan is required in PSIR to correct the phase of the signal and recover its true polarity (positive or negative), allowing tissues with different T1 times to be accurately distinguished — especially around the null point where magnitude reconstruction fails.

Property	Description
Sign preservation	Negative Mz remains negative after reconstruction.
TI insensitivity	Nulling is not strictly required; negative tissues remain visible.
Better scar–myocardium contrast	Particularly in late gadolinium enhancement imaging.
Binary phase	Only 0 or π; fine phase information is discarded after polarity decision.
Reference image required	Without reference, true PSIR cannot be done.

PSIR in CMR

- Magnitude vs PSIR Reconstruction in CMR
  - Magnitude Reconstruction
    - Ignores signal polarity.
    - Nulled tissue → dark (0 signal).
    - Negative magnetization folds up → appears bright.
    - Contrast depends strongly on TI selection.
  - PSIR
    - Preserves signal sign, displaying nulled tissue as gray.
    - Shorter T1 → brighter; longer T1 → darker.
    - Contrast preserved even if TI varies.
    - Enables robust myocardium nulling, better blood–scar contrast, less TI sensitivity, and improved reproducibility.
- Clinical Impact
  - Normal myocardium (negative magnetization) appears black, improving scar detection.
  - Less sensitive to T1 variability across lesions.
  - PSIR reduces artifacts and improves image consistency, especially with arrhythmias.
- Trade-off: PSIR needs two sets of data for each image doubled scan time and potential beat-to-beat mismatch.

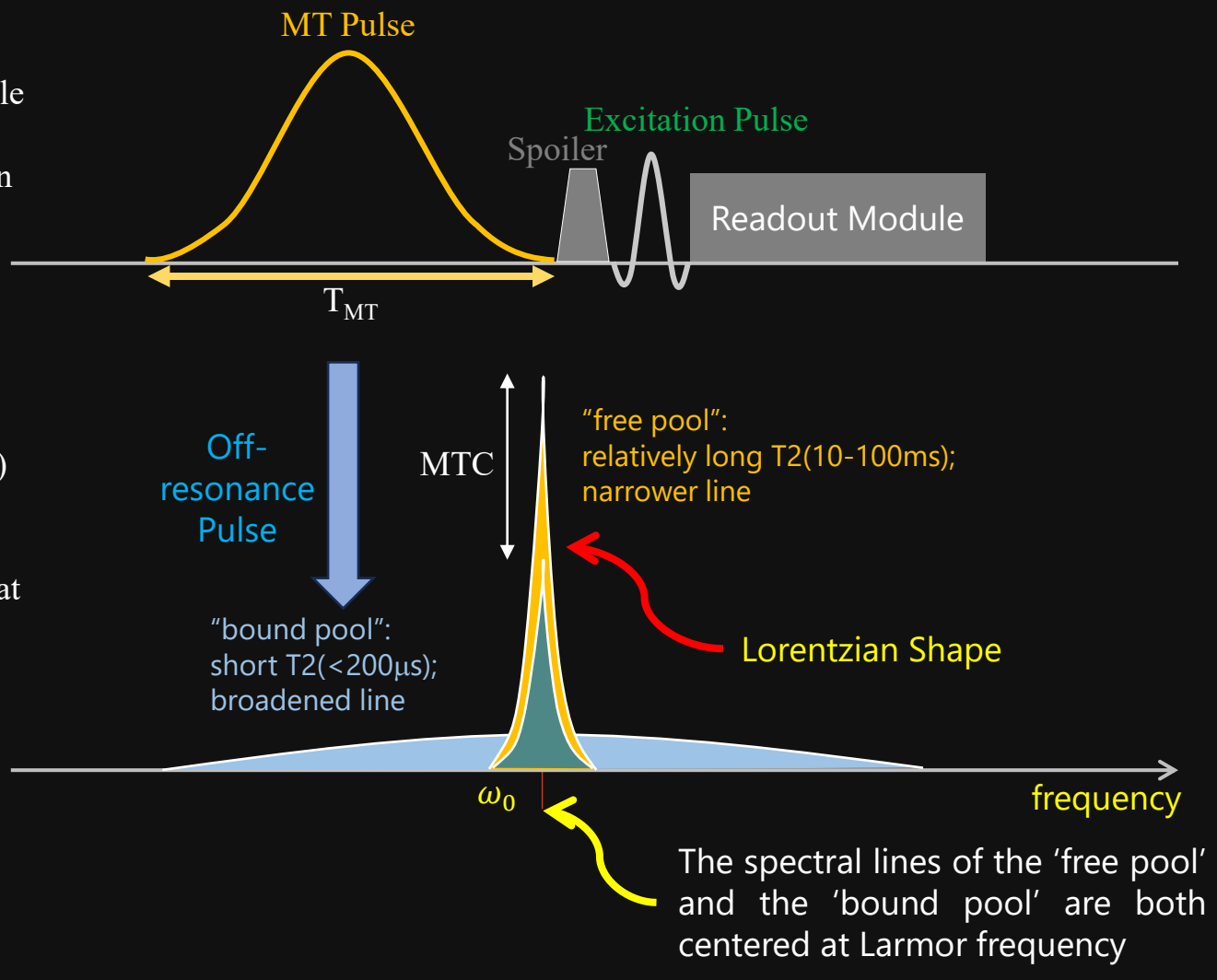




Simulation parameters at 1.5T:
Myocardium (native): ~950 ms
Blood (native): ~1500 ms
Scar (post MI): ~450 ms
CSF: ~4000 ms
Post-contrast (typical 10–15 min after GBCA):
Myocardium: ~450 ms
Blood: ~300 ms
Scar: ~250 ms
CSF: ~2500 ms (minimal enhancement)
TI = 300 ms (typical late gadolinium enhancement)

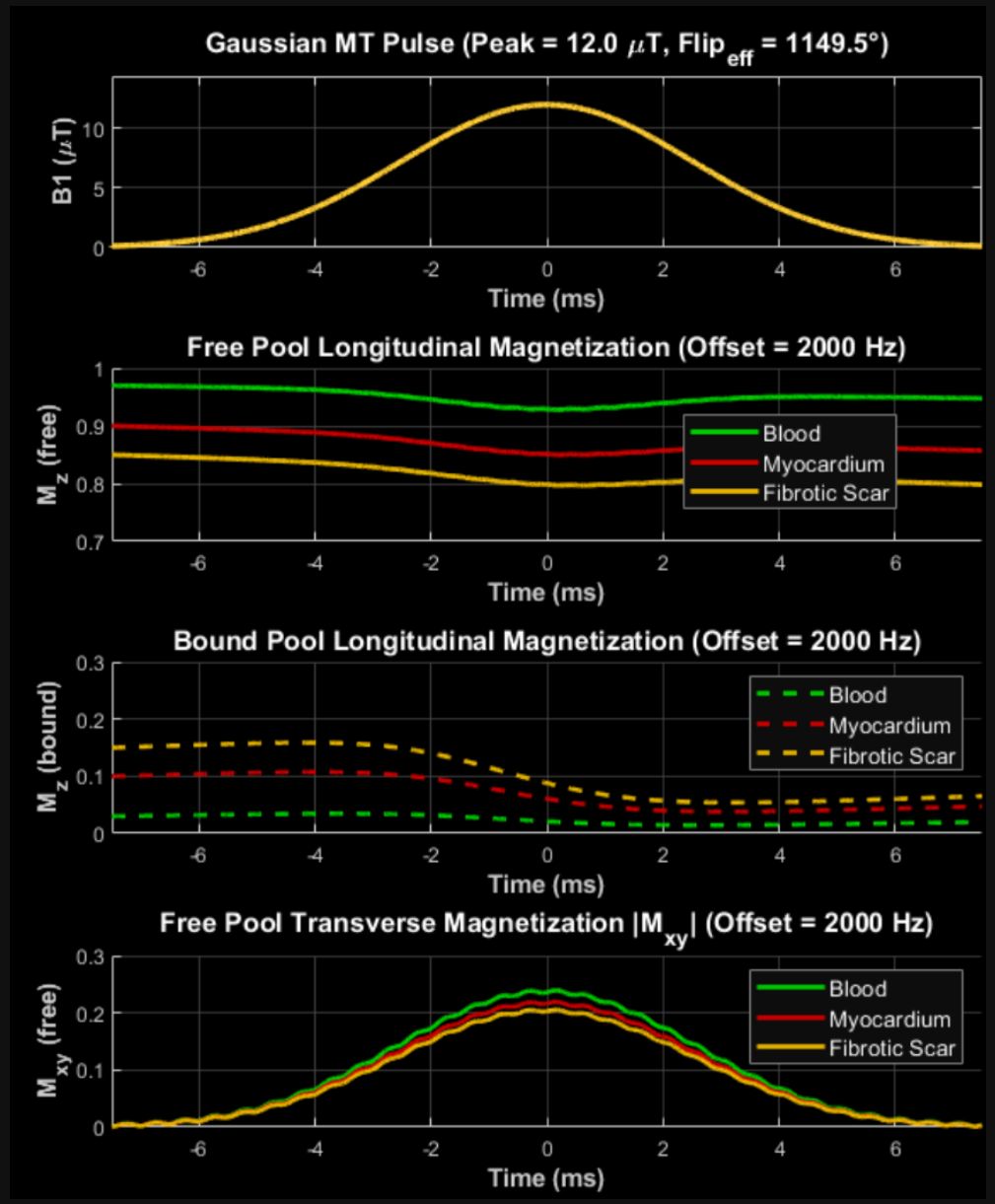
MTC Module

- MT pulses are off-resonance saturation pulses designed to selectively saturate the bound proton pool (macromolecules) while minimizing direct saturation of the free water pool. The resulting cross-relaxation lowers the free pool's longitudinal magnetization → MT contrast. In CMR, MT pulse is used to suppress myocardium signal more than blood, improving contrast and border definition.
- MTC Preparation:
  - Off-resonance RF pulses (500–1500 Hz from water resonance)
  - Flip angle: 500°–1000°, duration: 5–20 ms (single or train)
  - Often followed by a spoiler to remove residual transverse magnetization
  - Applied before imaging readout, similar to T2prep or fat sat modules.
- Common configurations:
  - Gaussian or sinc-shaped RF pulses
  - Continuous wave (CW) or pulsed MT
  - Frequency offset ~1–2 kHz below water resonance
- MTC is often combined with other prep modules:
  - T2prep + MTC → black-blood imaging
  - IR + MTC → improved LGE border sharpness
  - bSSFP + MTC → brighter blood pool



1. Wolff SD, Balaban RS. Magnetization transfer contrast (MTC) and tissue water proton relaxation in vivo. Magn Reson Med. 1989 Apr;10(1):135-44. doi: 10.1002/mrm.1910100113. PMID: 2547135.
2. Henkelman RM, Stanisz GJ, Graham SJ. Magnetization transfer in MRI: a review. NMR Biomed. 2001 Apr;14(2):57-64. doi: 10.1002/nbm.683. PMID: 11320533.
3. Weber OM, Speier P, Scheffler K, Bieri O. Assessment of magnetization transfer effects in myocardial tissue using balanced steady-state free precession (bSSFP) cine MRI. Magn Reson Med. 2009 Sep;62(3):699-705. doi: 10.1002/mrm.22053. PMID: 19572387.
4. Holtackers, R.J., Van De Heyning, C.M., Chiribiri, A. *et al.* Dark-blood late gadolinium enhancement cardiovascular magnetic resonance for improved detection of subendocardial scar: a review of current techniques. *J Cardiovasc Magn Reson* **23**, 96 (2021). https://doi.org/10.1186/s12968-021-00777-6

MTC Module – How does it work?



- MT effect on signal

- $S_{MT} = S_0 \cdot (1 - f_{MT}) = kM_0 sina \cdot (1 - f_{MT})$
- Where f_{MT} depends on 1) Bound pool fraction (higher in myocardium than blood); 2) RF power and offset; 3) T1 recovery during prep
- Myocardium has abundant macromolecules → stronger MT effect → greater signal attenuation.
- Blood has few macromolecules → minimal MT effect → signal is preserved.
- This difference in MT saturation enhances blood–myocardium contrast even when other parameters (e.g., T1) are similar.

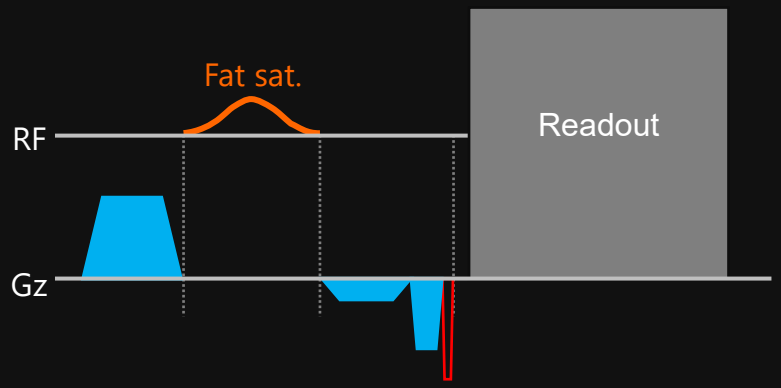
- MT contrast ratio can be roughly expressed as:

- $MTR = \frac{S_0 - S_{MT}}{S_0}$
- In CMR, $MT\ contrast\ ratio \approx \frac{S_{blood}}{S_{myocardium}}$
- Typical MT-based myocardium–blood contrast ratios in cardiac imaging at 1.5–3 T are approximately 1.5–2.0, meaning blood signal intensity can be up to twice as high as myocardium on balanced SSFP without contrast agents.

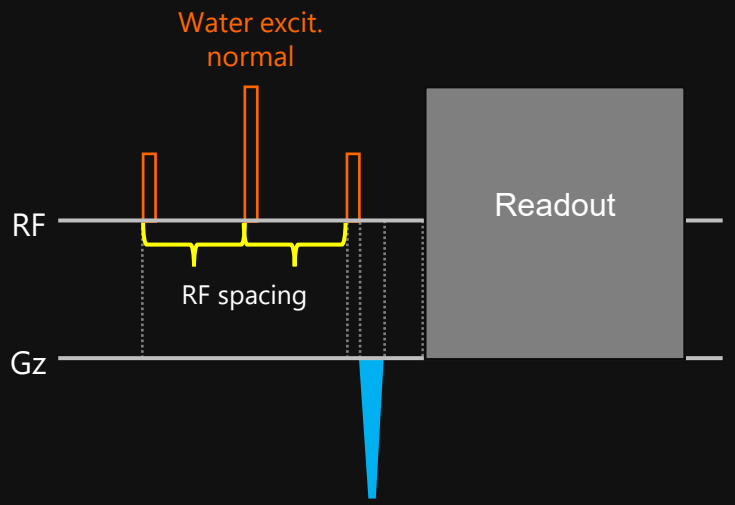
Tissue	MTR (1.5 T)	MTR (3.0 T)
Blood	0.05 – 0.10	0.10 – 0.15
Myocardium	0.25 – 0.35	0.30 – 0.40
Fibrotic scar	0.35 – 0.45	0.40 – 0.50
White matter	0.40 – 0.50	0.45 – 0.55
Gray matter	0.30 – 0.40	0.35 – 0.45
Skeletal muscle	0.20 – 0.30	0.25 – 0.35
Fat / adipose	~0.05	~0.05
CSF	<0.02	<0.02

Fat Suppression Module

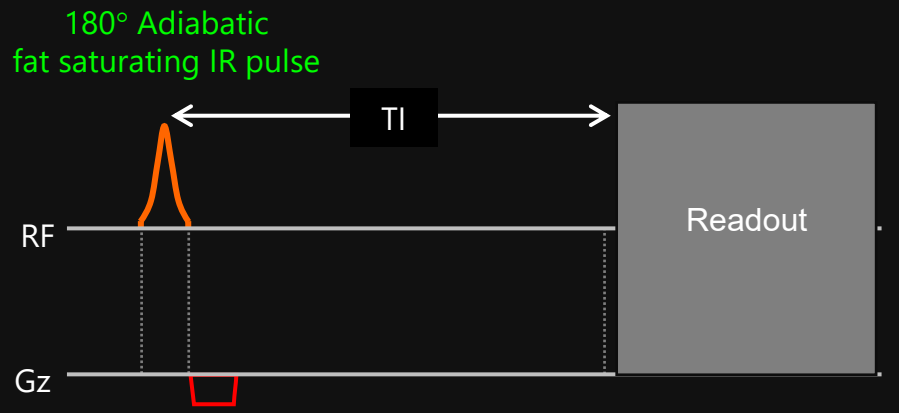
Fat sat.: Suppresses the fat signal and has no effect on TE while TR might be prolonged. This extends the measurement time as well.



Water excit. (Normal 121 or Fast 11): Suppresses the fat signal, effects moderate extension of TE and TR.



SPAIR: Prior to image saturation, the fat signal is suppressed via a frequency-selective adiabatic inversion pulse.

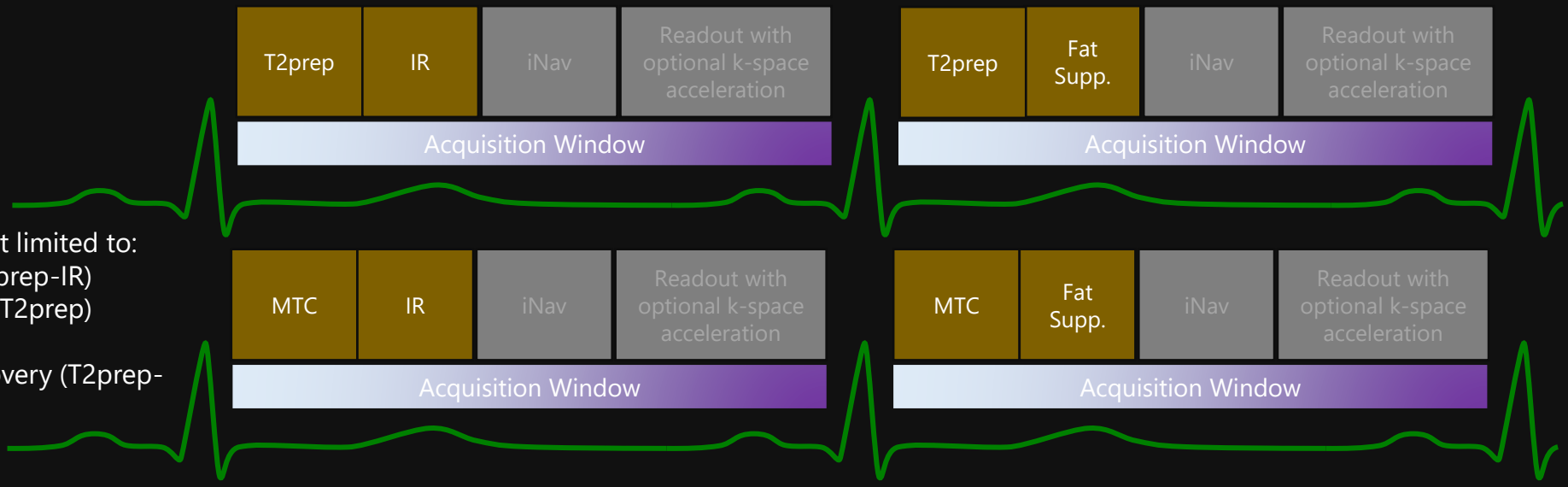


Hybrid MP Modules: Further Optimizing Tissue and Blood Contrast

- Hybrid magnetization-preparation (MP) modules are often utilized to enhance tissue contrast, suppress unwanted signals, and improve imaging efficiency. They play a key role in LGE imaging, whole-heart angiography, and edema detection. They are frequently implemented in free-breathing acquisitions with motion correction, improving image quality and patient comfort.

- By combining elements such as T2prep, IR, PSIR, and MTC, these modules can:
  - Achieve flow-independent black-blood contrast
  - Enable simultaneous nulling of blood and myocardium
  - Provide robust fat suppression

- Common hybrid MP modules include, but not limited to:
  - Hybrid T2prep-Inversion Recovery (T2prep-IR)
  - Hybrid Inversion Recovery-T2prep (IR-T2prep)
  - Hybrid MTC-IR / MTC-FatSat
  - T2prep Phase-Sensitive Inversion Recovery (T2prep-PSIR)
  - …

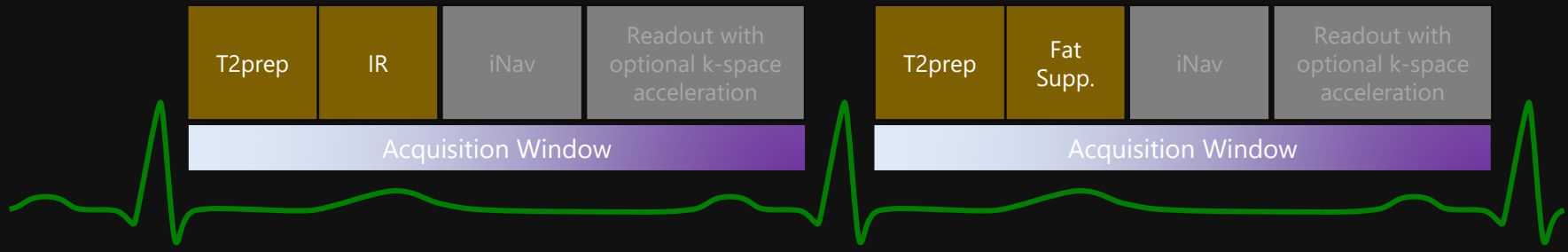


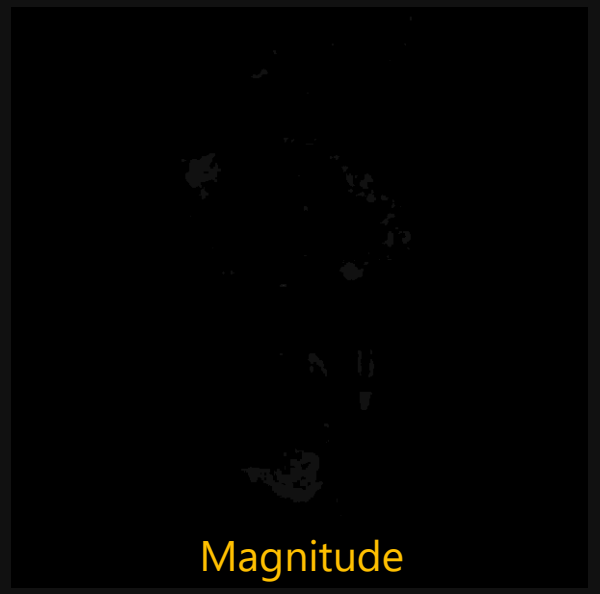
Hybrid MP Modules In CMR: T2prep+IR, T2Prep+Fat Supp. in BOOST

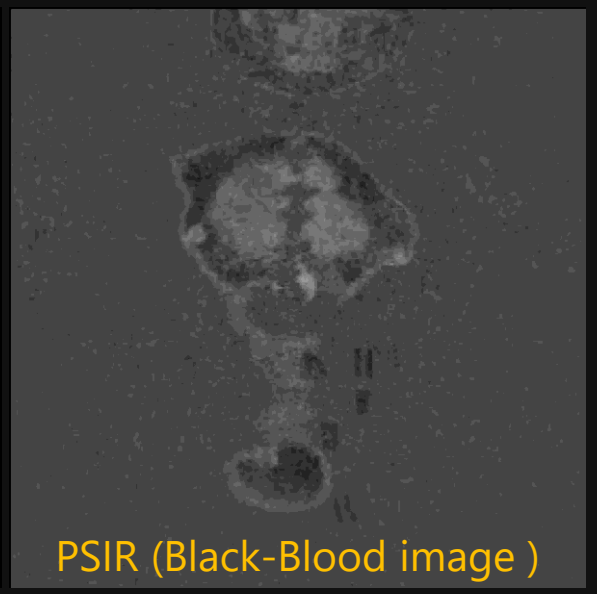
- BOOST (Bright-blood and black-blOOd phase SensiTive) is a 3D whole-heart MRI technique designed for simultaneous noncontrast-enhanced visualization of the coronary lumen, myocardium, and thrombus/hemorrhage.
- It alternates the acquisition of two bright-blood datasets with different preparatory pulses to vary blood/myocardium contrast. These datasets are combined in a phase-sensitive inversion recovery (PSIR)-like reconstruction to produce a coregistered black-blood dataset.

A T2Prep-IR module is applied at odd heartbeats (a) to enable coronary lumen visualization with improved contrast between blood and myocardium. A short TI is set to null the signal from epicardial fat.

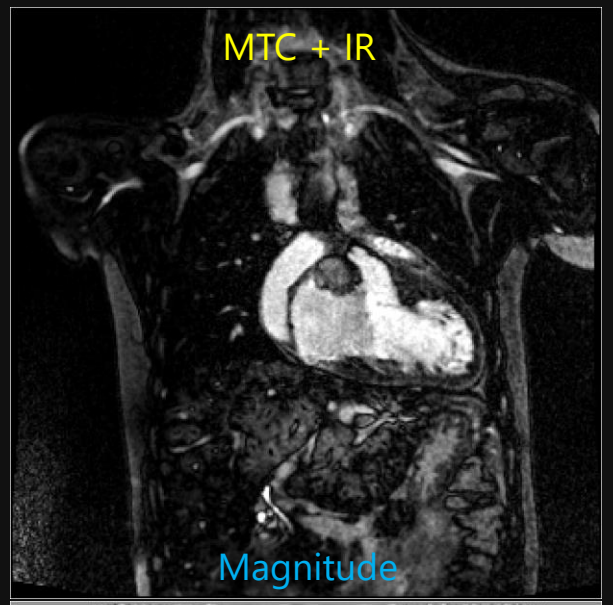
At even heartbeats (b), data acquisition is T2 prepared and performed with a high flip angle using spectral presaturation for fat suppression.

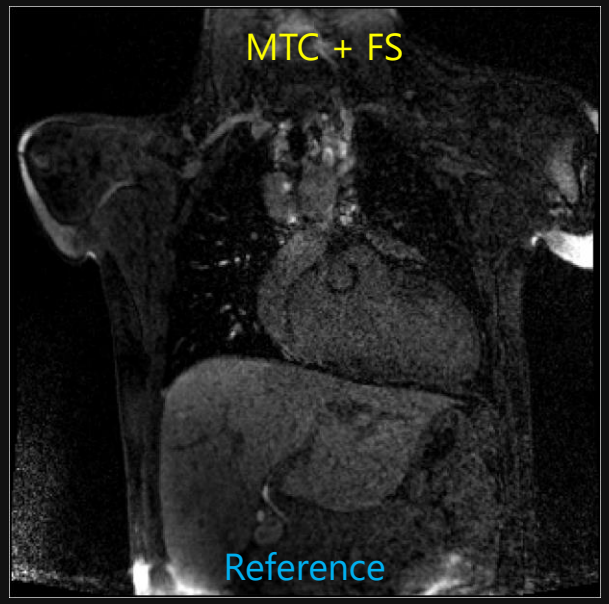


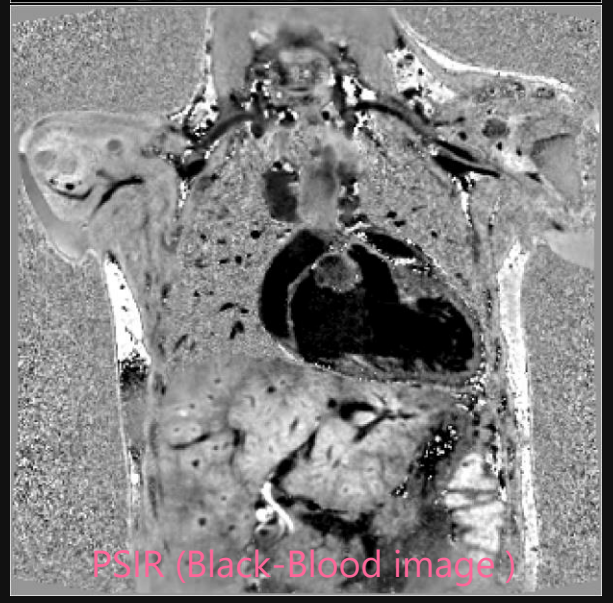




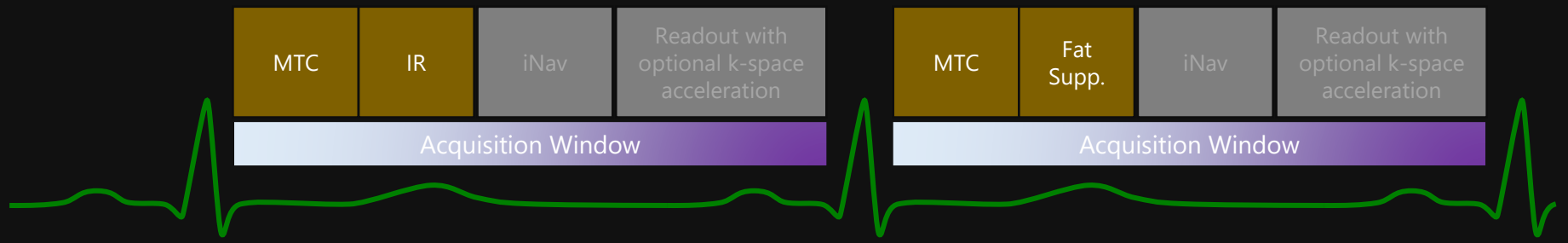
Hybrid MP Modules In CMR: MTC+IR, MTC+Fat Supp. in BOOST







MTC+IR Hybrid Module: MTC saturates bound macromolecular protons to suppress background tissue, while IR selectively nulls tissues based on T1. Together, they enhance myocardium-to-blood and vessel wall contrast, providing robust blood and tissue differentiation in CMR.



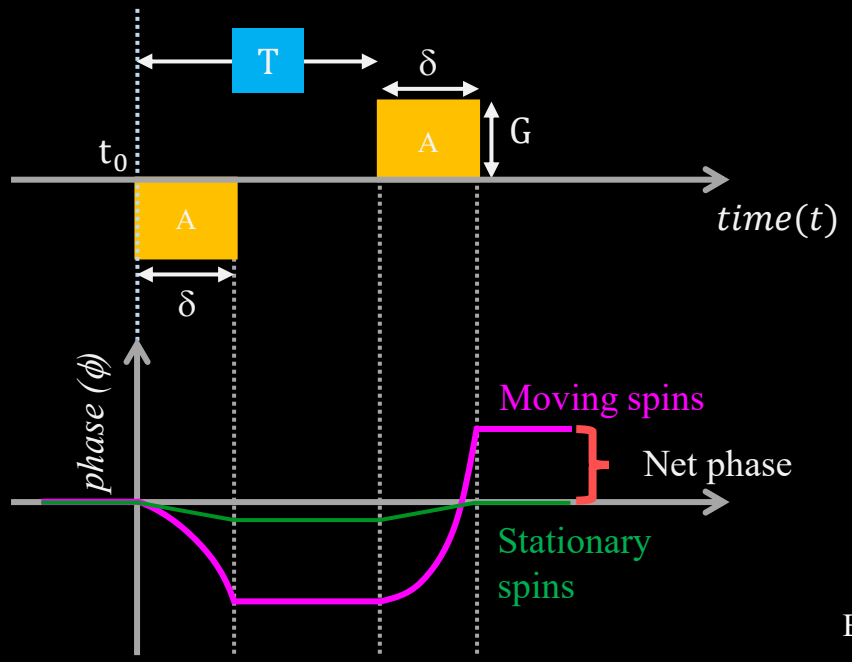
Blood Flow and MRI Contrast

- Flow enhances contrast by differentiating moving blood from static tissues.
  - Time-of-Flight (TOF) Effect
    - Fresh inflowing blood remains unsaturated → bright signal.
    - Stationary myocardium becomes partially saturated → darker background.
  - Phase Contrast (PC) Effect
    - Moving spins accumulate velocity-dependent phase shifts.
    - Enables direct flow quantification in magnitude and phase images.
    - Foundation of 2D PC-MRI and 4D Flow MRI.
  - 4D Flow MRI
    - Provides time-resolved 3D velocity information.
    - Allows calculation of flow volume, velocity vectors, WSS, and pressure gradients.
    - Supports comprehensive hemodynamic assessment.
  - Flow Void
    - Rapid/turbulent flow → intravoxel dephasing → signal loss (dark lumen).
- Blood flow generates both image contrast and quantitative velocity information, enabling detailed visualization and measurement of complex cardiovascular hemodynamics.

Bipolar Gradient, Velocity Encoding, and Phase contrast MRI

Bipolar Gradient Waveform

$$G(t) = \begin{cases} -G & 0 \le t < \delta \\ 0 & \delta \le t \\ +G & T \le t < T + \delta \\ 0 & t > T + \delta \end{cases}$$



Bipolar Gradient Moment Calculations

$M_0 = \int_0^t G(\tau)d\tau = -G\delta + G\delta = 0$

$M_1 = \int_0^t \tau G(\tau)d\tau = -G\int_0^\delta t dt + G\int_T^{T+\delta} t dt = -G\cdot\frac{1}{2}\delta^2 + G\cdot\frac{1}{2}[(T+\delta)^2 - T^2] = G\delta T$

- the velocity-encoding term in PC-MRI

$M_2 = \int_0^t \tau^2 G(\tau)d\tau = -G\int_0^\delta t^2 dt + G\int_T^{T+\delta} t^2 dt = -G\cdot\frac{1}{3}\delta^3 + G\cdot\frac{1}{3}[(T+\delta)^3 - T^3] = G\delta(T^2 + \delta T)$

- represents acceleration sensitivity, which is usually negligible in standard PC-MRI.

$$\phi(t) = \phi_0 + \gamma\int_0^t G(\tau)\cdot r(\tau)d\tau \xrightarrow{r(\tau)=r_0+v\tau+\frac{1}{2}a\tau^2}$$

$$= \phi_0 + \gamma\int_0^t G(\tau)\cdot(r_0 + v\tau + \frac{1}{2}a\tau^2)d\tau$$

$$= \phi_0 + \gamma\left[r_0\underbrace{\int_0^t G(\tau)d\tau}_{M_0} + v\underbrace{\int_0^t \tau G(\tau)d\tau}_{M_1} + \frac{1}{2}a\underbrace{\int_0^t \tau^2 G(\tau)d\tau}_{M_2}\right]$$

Background phase

M0: static phase (position dependent)

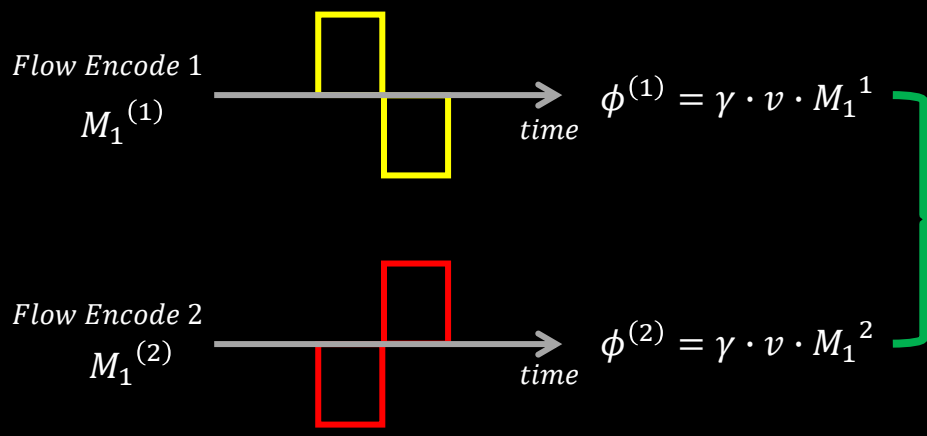
M1: velocity-encoding term

M2: acceleration-encoding term.

- Phase-contrast MRI (PC-MRI) refers to a family of MR imaging techniques that exploit the fact that spins moving through magnetic field gradients acquire a different phase than stationary spins, enabling the generation of images with controlled sensitivity to flow.
- The cornerstone of PC-MRI is the use of a bipolar gradient. Standard PC-MRI techniques are typically based on a gradient-echo acquisition.
- When stationary and moving spins are subjected to a pair of bipolar gradients, the resulting flow- or motion-dependent phase shifts can be used to encode the underlying blood-flow velocity directly into the MR signal phase. By measuring changes in phase, the velocity of moving spins can be quantitatively computed.

Velocity Encoding Schemes

Scheme: Difference in 1st moments ΔM_1



$$\phi^{(1)} = \gamma \cdot v \cdot {M_1}^1$$

$$\phi^{(2)} = \gamma \cdot v \cdot {M_1}^2$$

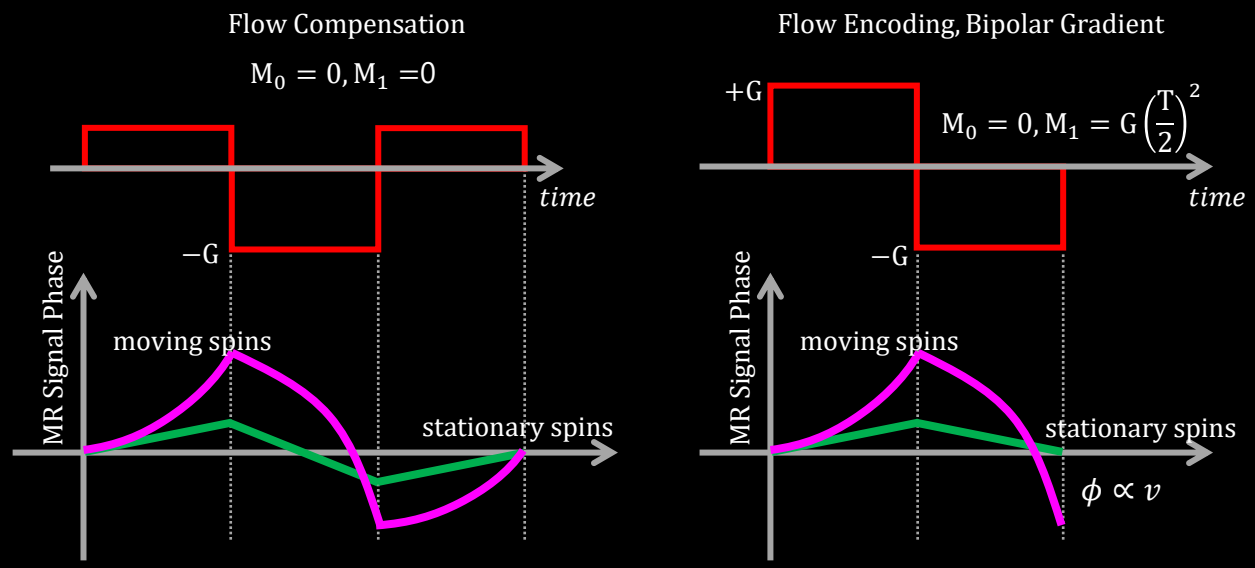
$$\Delta\phi = \phi^{(1)} - \phi^{(2)} = \gamma \cdot v \cdot \Delta M_1 \Rightarrow v = \frac{\Delta\phi}{\gamma \cdot \Delta M_1}$$

$$v = \frac{\Delta\phi}{\pi} \cdot v_{enc}$$

$$v_{enc} = \frac{\pi}{\gamma \cdot \Delta M_1}$$

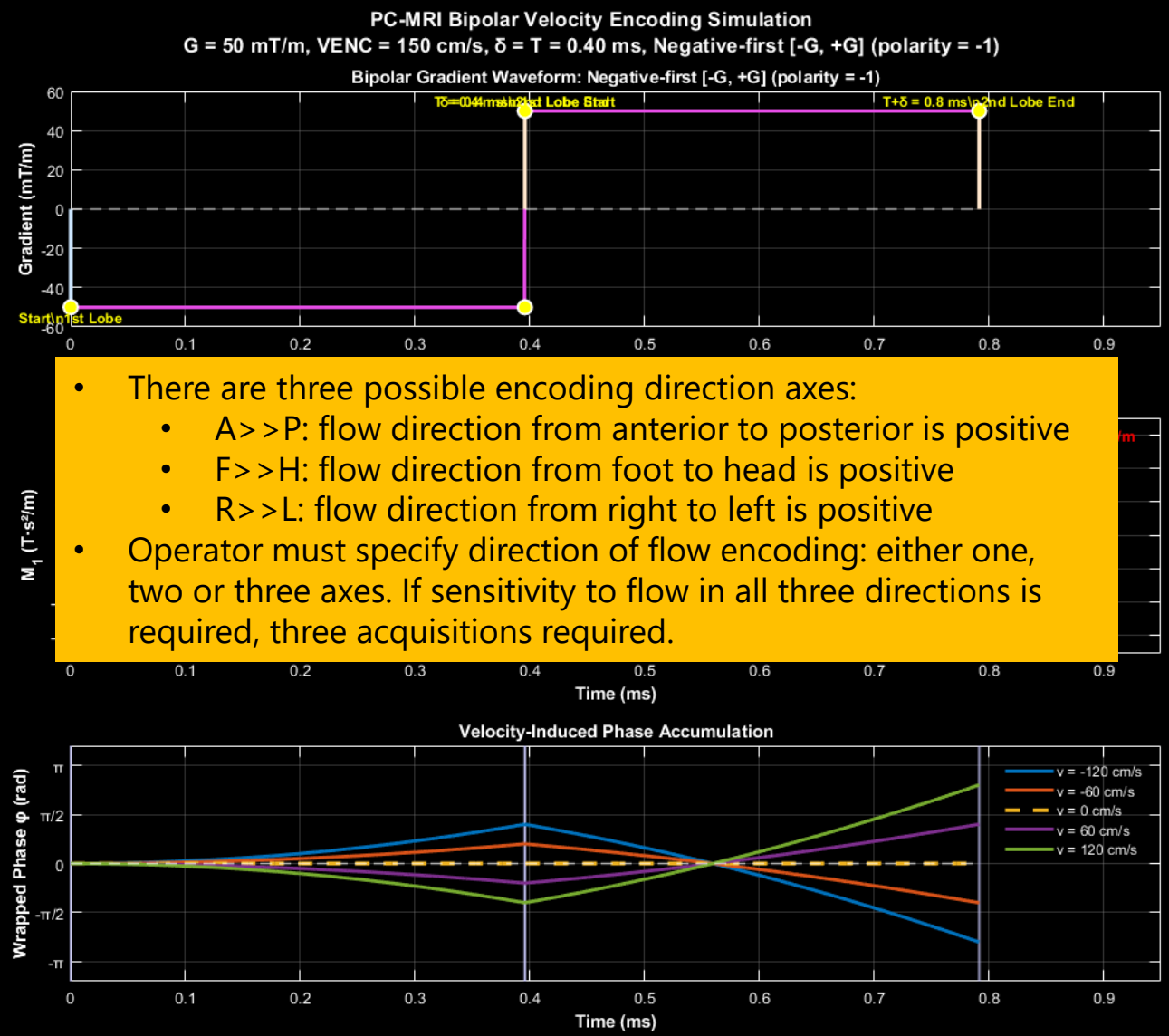
For any velocity-encoding scheme, the primary objective is to eliminate unwanted background phase variations (assuming the moving spins have negligible acceleration) while ensuring that the accumulated phase is linearly proportional to the spin velocity.

Scheme: Flow Compensation+Flow Encoding

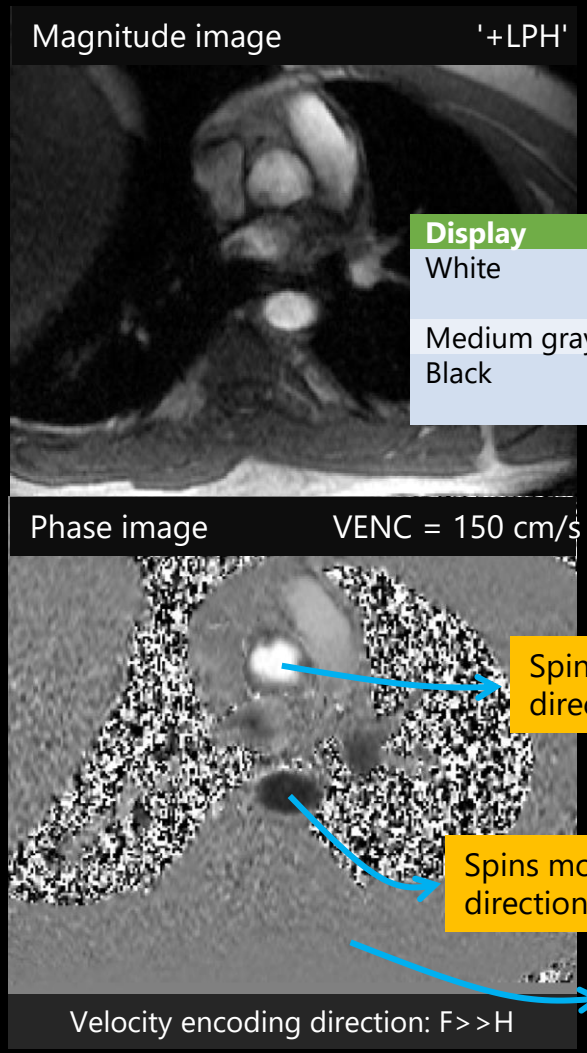


- Flow compensation is achieved when the first gradient moment is nulled at the echo time (M0=0, M1=0), ensuring that stationary and uniformly moving spins experience zero net phase shift, thereby minimizing motion-induced dephasing.
- In contrast, flow encoding intentionally introduces a nonzero first moment (M0=0, M1≠0) by applying a bipolar gradient pair—one lobe of positive polarity and one of negative polarity—so that stationary spins experience no net phase change while moving spins acquire a velocity-dependent phase shift.
- $\Delta\phi = \phi^{(1)} - \phi^{(2)} = 2\gamma \cdot v \cdot M_1$
- In practical PC-MRI sequence design, alternating between flow-compensated and flow-encoded gradient configurations enables quantitative velocity mapping while suppressing background phase contributions.

Velocity Encoding Direction



- There are three possible encoding direction axes:
  - A>>P: flow direction from anterior to posterior is positive
  - F>>H: flow direction from foot to head is positive
  - R>>L: flow direction from right to left is positive
- Operator must specify direction of flow encoding: either one, two or three axes. If sensitivity to flow in all three directions is required, three acquisitions required.



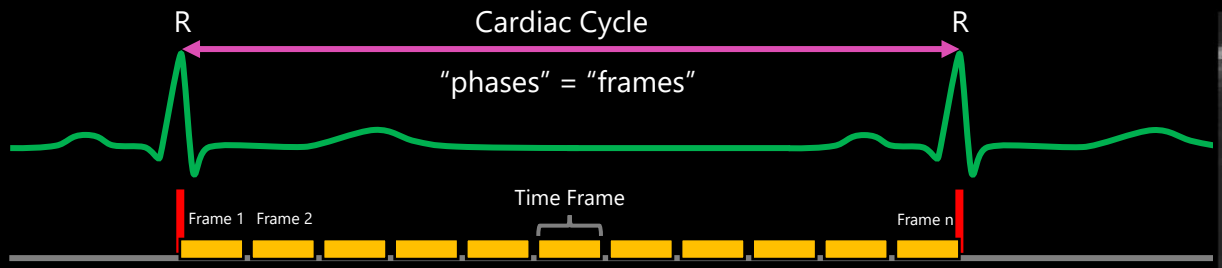
Display	Flow Velocity
White	($+v_{enc}$) Maximum flow velocity in positive direction
Medium gray	(zero) Stationary tissue (no flow)
Black	($-v_{enc}$) Maximum flow velocity in negative direction

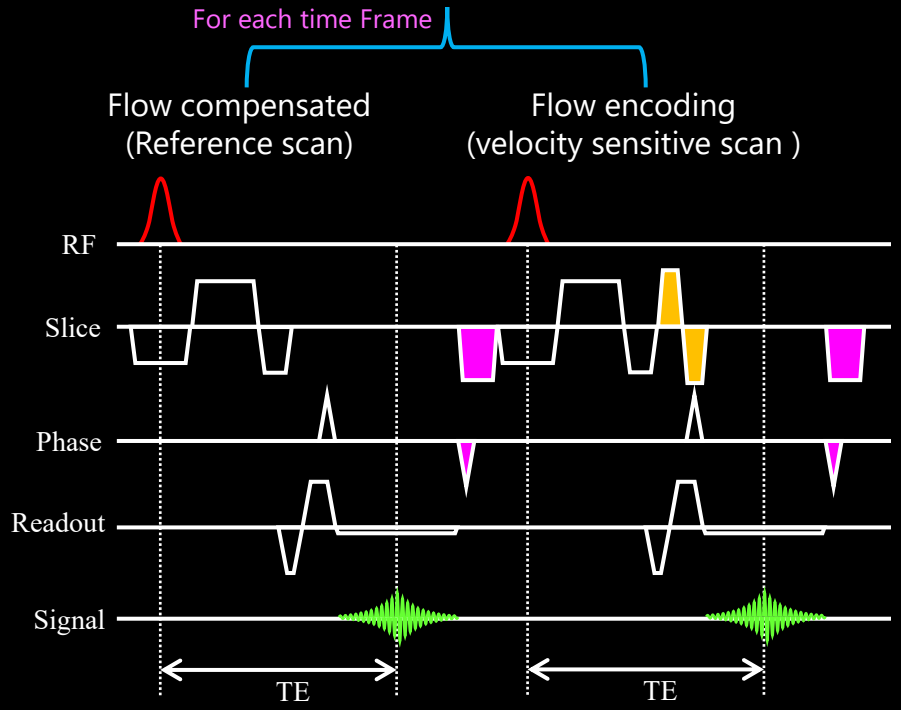
Spins moving in the same encoding direction (positive net phase shift)

Spins moving in the opposite encoding direction (negative net phase shift)

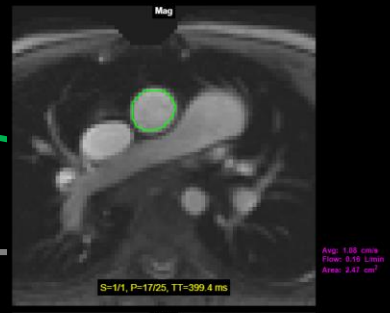
Stationary spins (net zero phase shift)

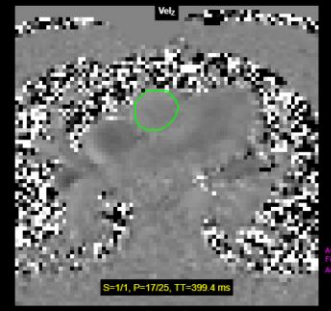
Standard 2D PC-MRI with Retrospective Gating for Flow Quantification

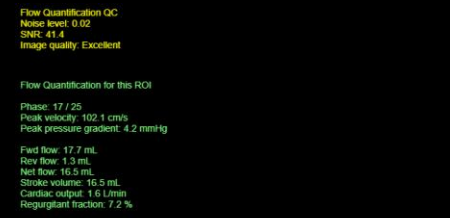




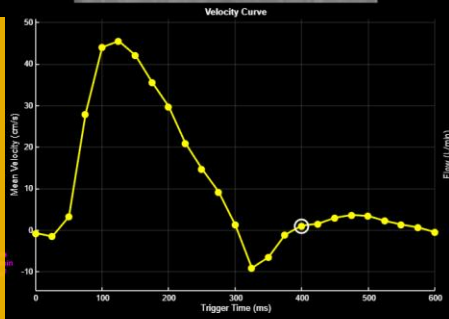
- Reference and velocity sensitive scan (added bipolar velocity encoding gradient) are acquired in direct succession.
- The subtraction of both data sets provides phase difference images which contain quantitative blood flow velocities.

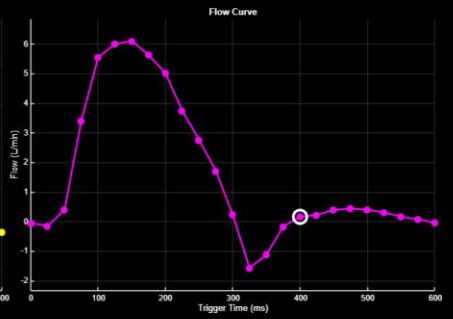


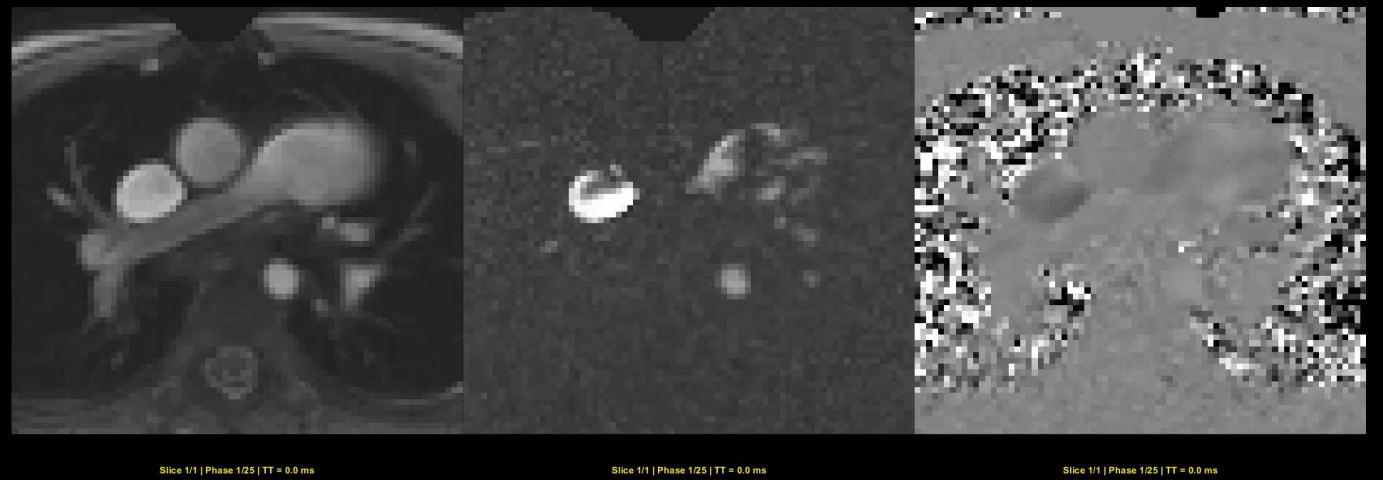




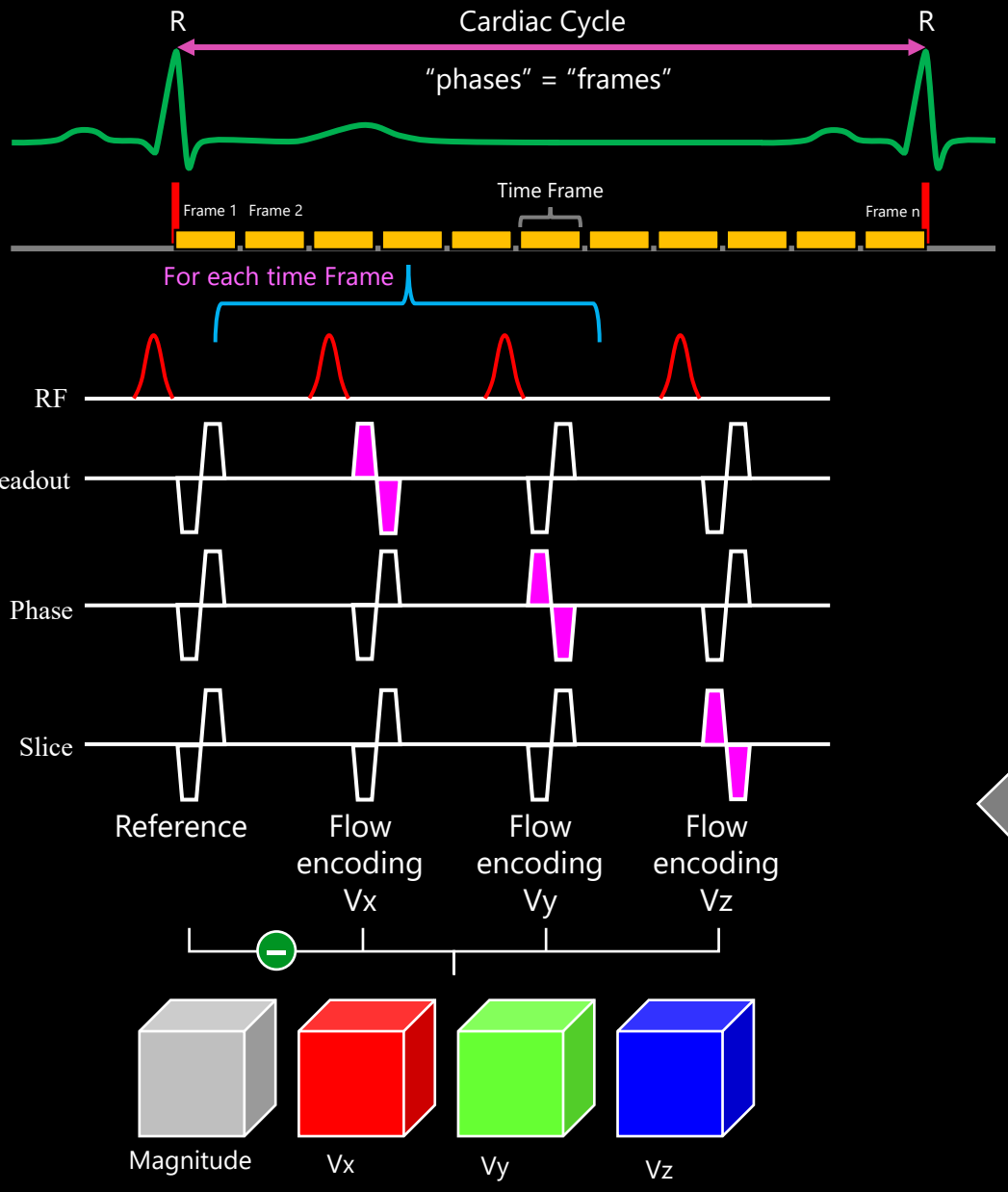
- For a given target voxel/ROI/vessel, the velocity changes during the cardiac cycle can be obtained by plotting the curve between time frame and the velocity measured in each frame.
- Based on the curve, the minimum, mean and peak velocities can be calculated.



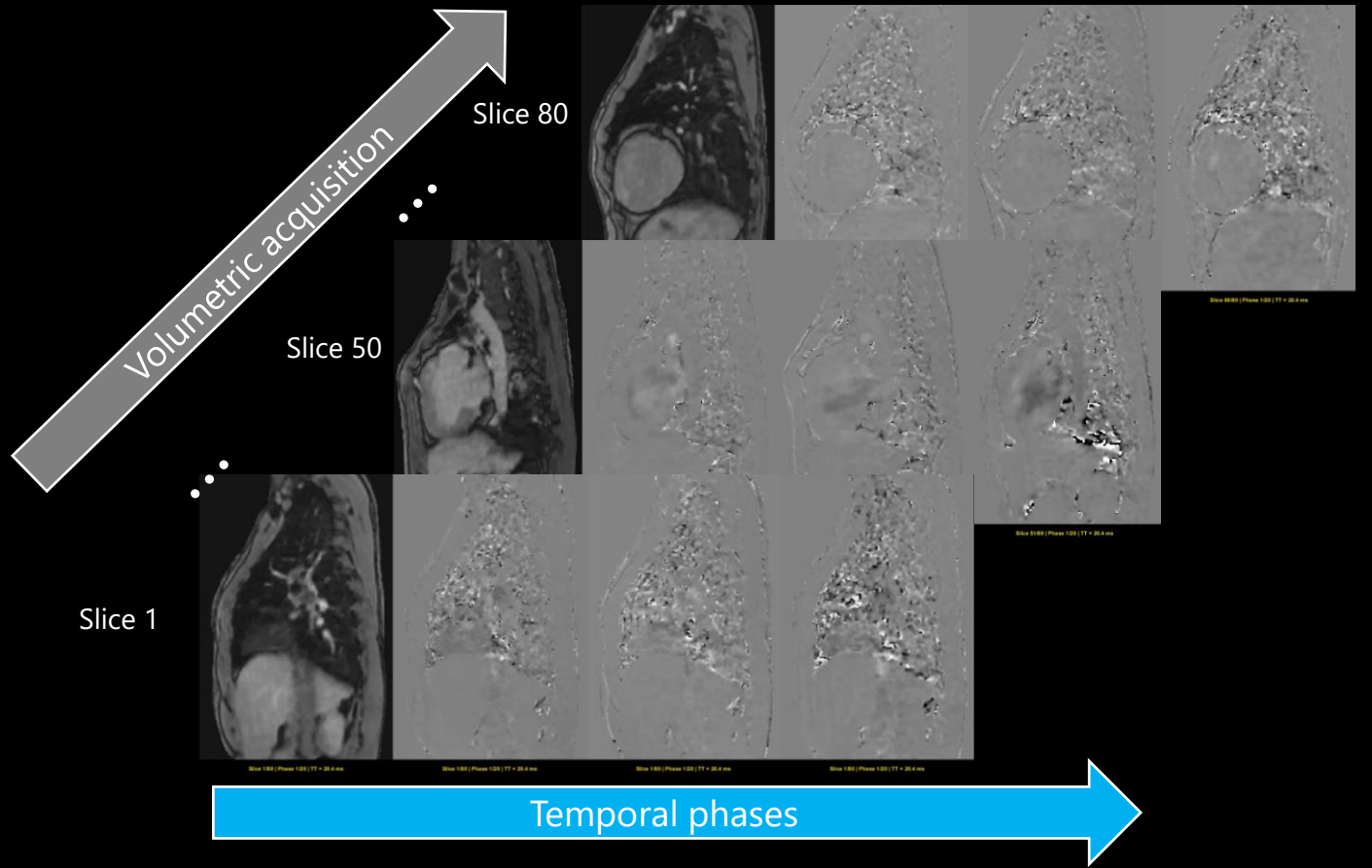




4D Flow MRI

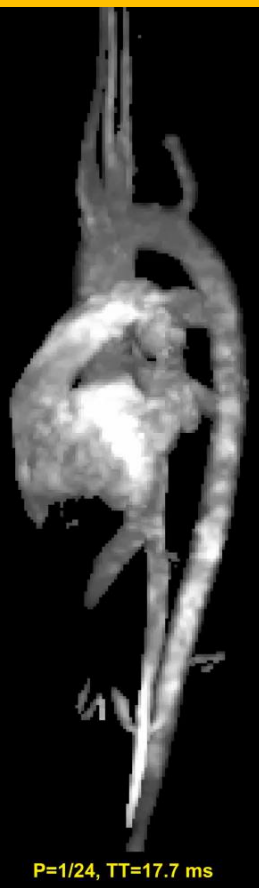


- By extending the velocity encoding schemes into three axes, the visualization of flow in multiple orientations is feasible, that's the basics for 4D flow MRI.
- 4D Flow MRI, also known as time-resolved 3D phase-contrast MRI, refers to volumetric acquisition of phase-contrast CMR with flow-encoding applied in all three spatial directions (x, y, z-axes) and to the dimension of time along the cardiac cycle (3D + time = 4D).
  - velocity in the x-direction, Vx,
  - velocity in the y-direction, Vy,
  - velocity in the z-direction, Vz,
  - time, i.e. time-frame in the cardiac cycle, or cardiac phase.

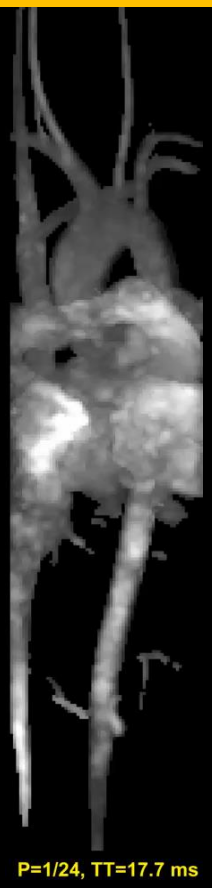


4D Flow MRI for Time-Resolved 3D Visualization and Quantification of Complex Hemodynamics

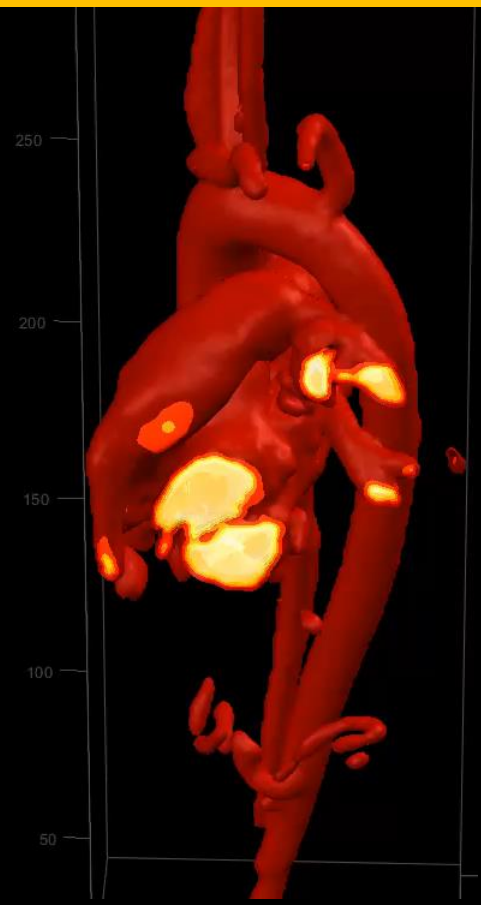
4D flow MRI provides comprehensive, time-resolved 3D assessment of cardiovascular hemodynamics, facilitating detailed visualization and quantification of complex blood flow dynamics.



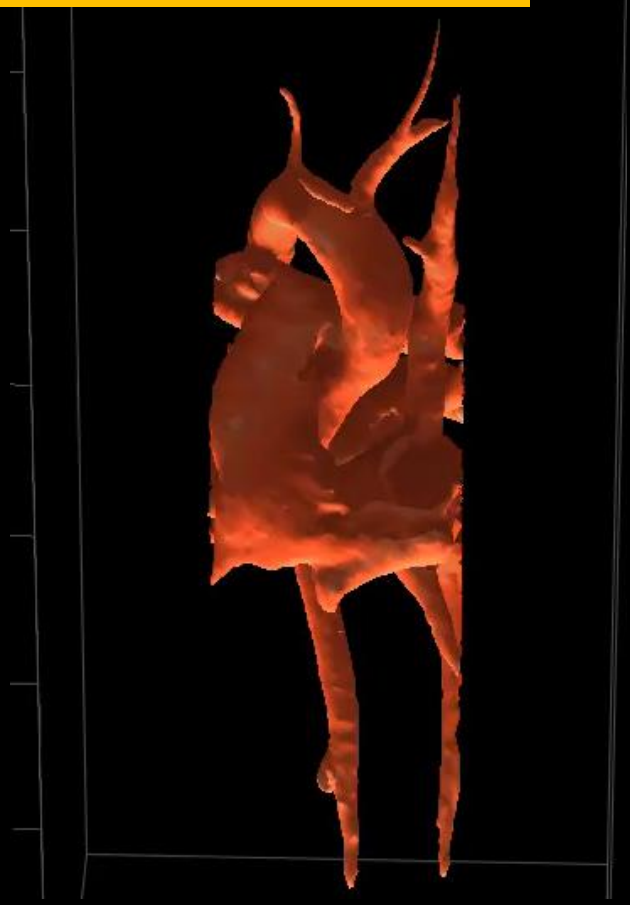
Time-resolved PCMRA (sagittal)



Time-resolved PCMRA (coronal)



Volume Rendering (VR)

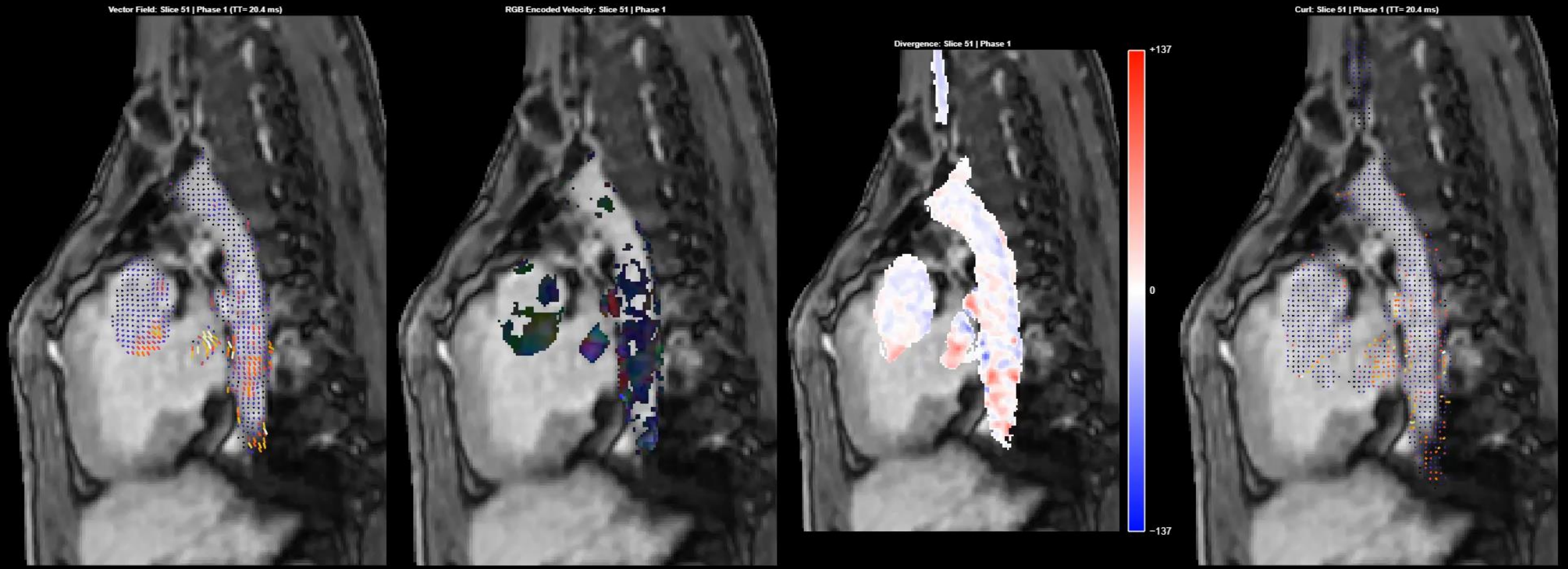


Shaded Surface Display (SSD)

Post-processing software is available at: https://github.com/sljzzw/mirQFlow

4D Flow MRI for Time-Resolved 3D Visualization and Quantification of Complex Hemodynamics

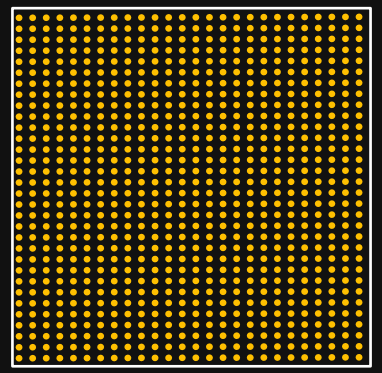
More complex flow patterns can be captured and analyzed through advanced imaging and post-processing techniques



Post-processing software is available at: https://github.com/sljzzw/mirQFlow

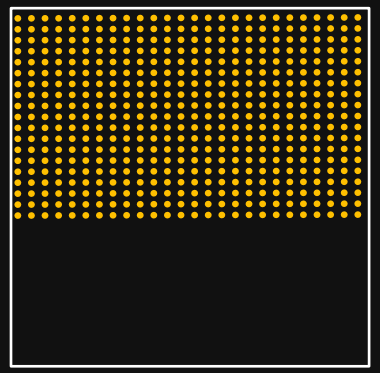
MRI Data Acceleration: Common 2D K-space undersampling strategies

Full k-space sampling: involves the dense collection of all data points in k-space.



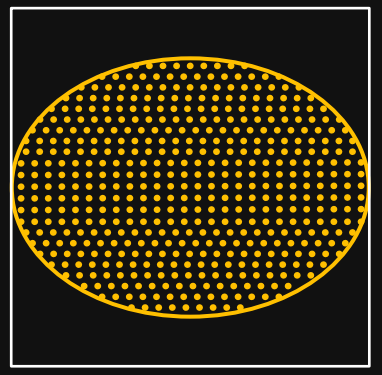
Full Sampling

Partial Fourier: only collects a portion of k-space and relies on the k-space Hermitian symmetry to reconstruct the missing information

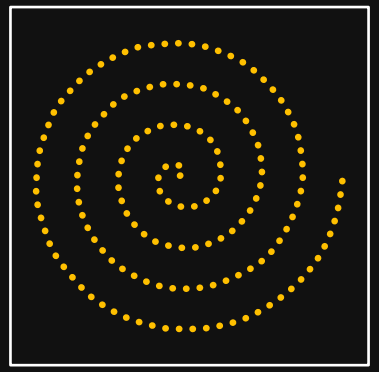


Partial Fourier

Elliptical Scanning: allows for the exclusion of data points in the corners of k-space that contribute minimally to image resolution, saving time without sacrificing resolution.



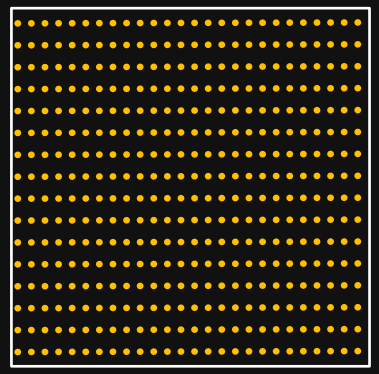
Elliptical Scanning



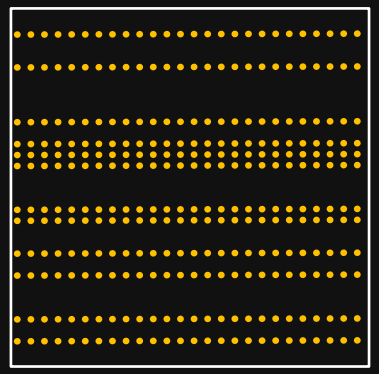
Spiral Sampling

Spiral Sampling: effective k-space coverage, higher SNR, inherent refocusing of motion and flow-induced phase error, which may reduce CSF flow artifacts.

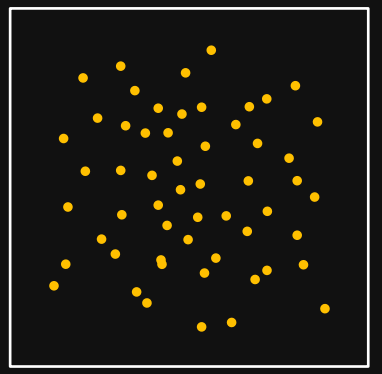
Parallel Imaging: utilize multiple receiver coils to acquire data simultaneously



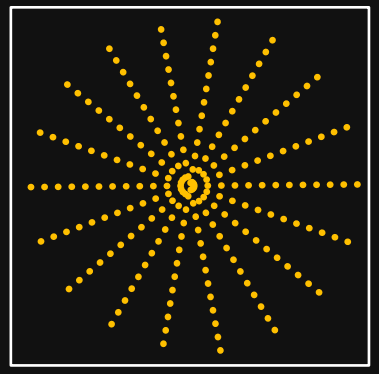
Regular undersampling



Pseudorandom undersampling



Random undersampling



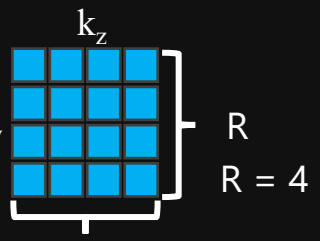
Radial Sampling

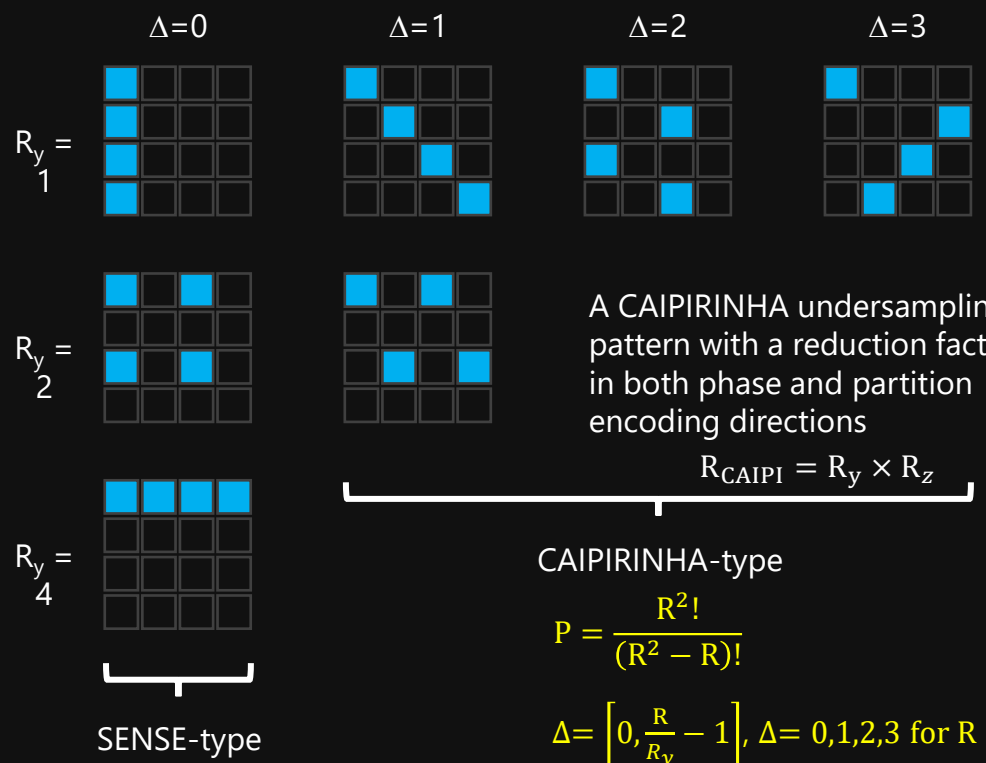
Radial Sampling: acquires data along radial lines emanating from the center of k-space. This approach naturally undersamples k-space, and the center of k-space contains the most critical image information

Compressed Sensing: allows for sparse sampling of k-space by leveraging the inherent sparsity of MR images in specific domains. Random undersampling involves randomly selecting a subset of k-space lines for acquisition.

Controlled Aliasing in Parallel Imaging Results in Higher Acceleration (CAIPIRINHA)

R×R elementary cell with R sampling positions to fill.





A CAIPIRINHA undersampling pattern with a reduction factor of 2 in both phase and partition encoding directions

$$R_{CAIPI} = R_y \times R_z$$

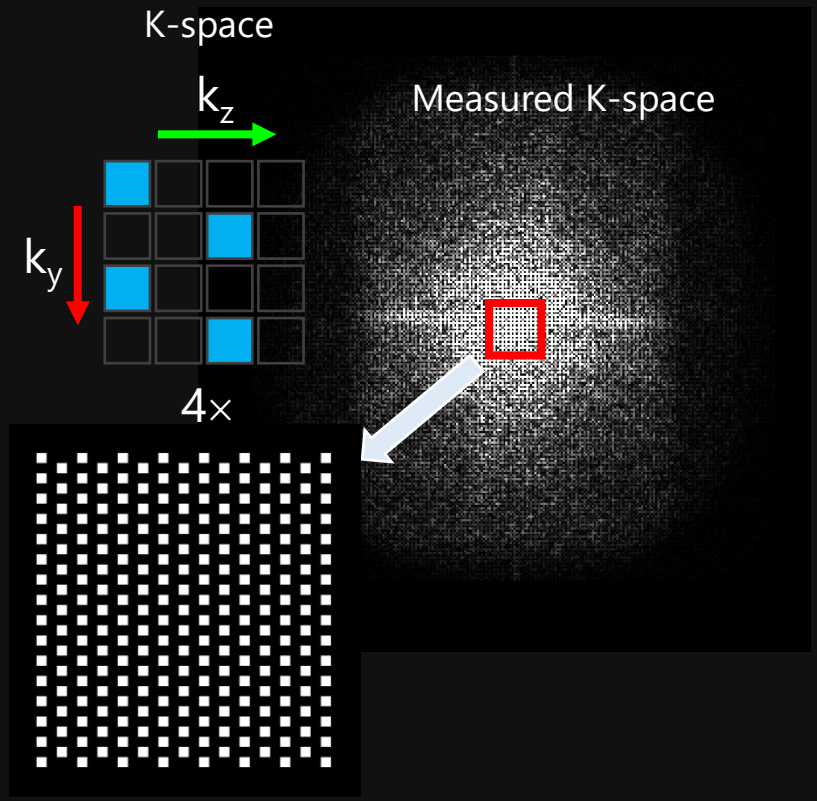
CAIPIRINHA-type

$$P = \frac{R^2!}{(R^2 - R)!}$$

$$\Delta = \left[0, \frac{R}{R_y} - 1\right], \Delta = 0,1,2,3 \text{ for } R = 4$$

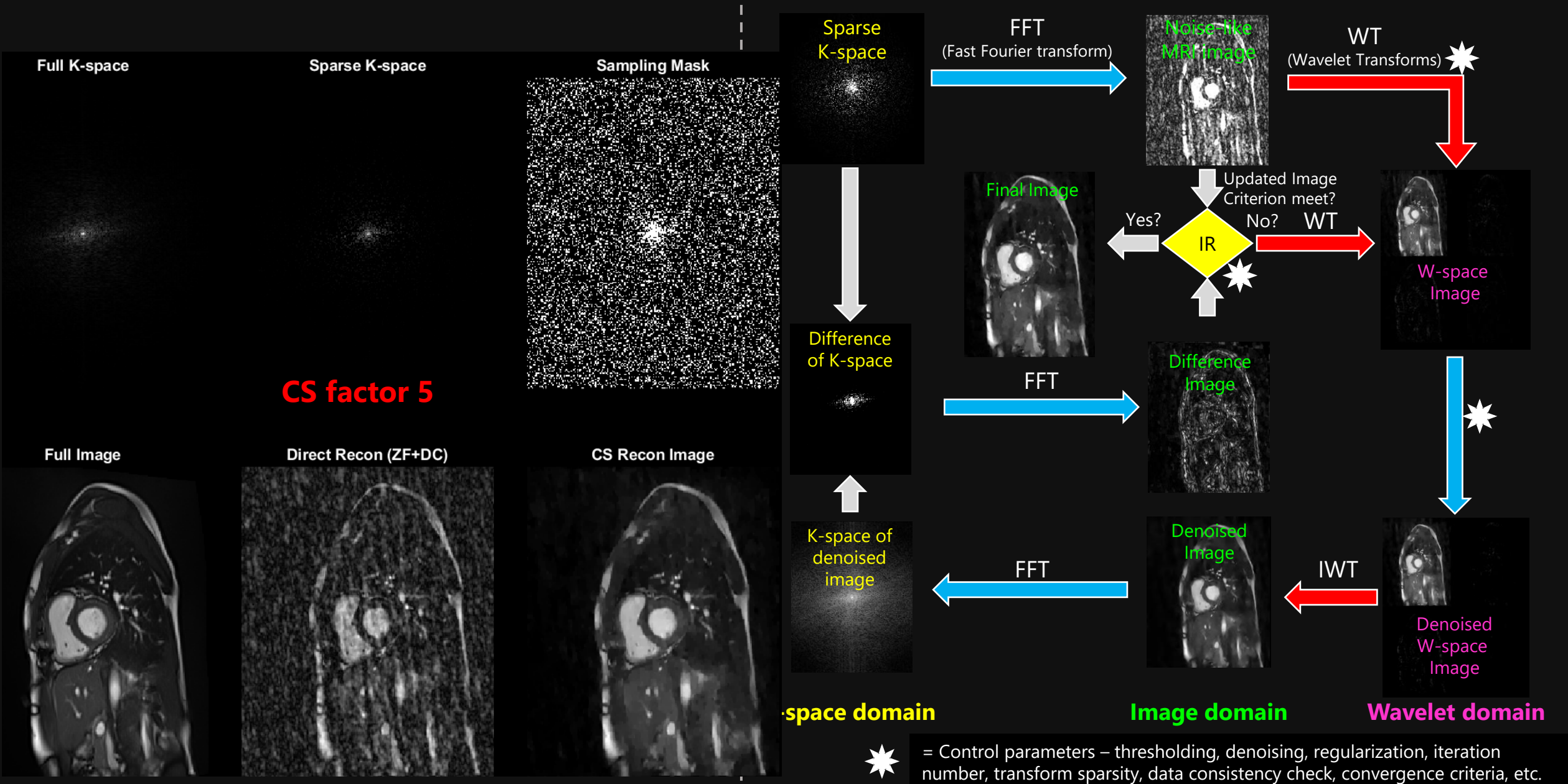
Note that only patterns starting at sampling position (1,1) in the binary sampling cell are considered. This arrangement leads to a reduced number of acceptable undersampled 2D sampling patterns

Optimized CAIPIRINHA Sampling Patterns in Volumetric Parallel Imaging



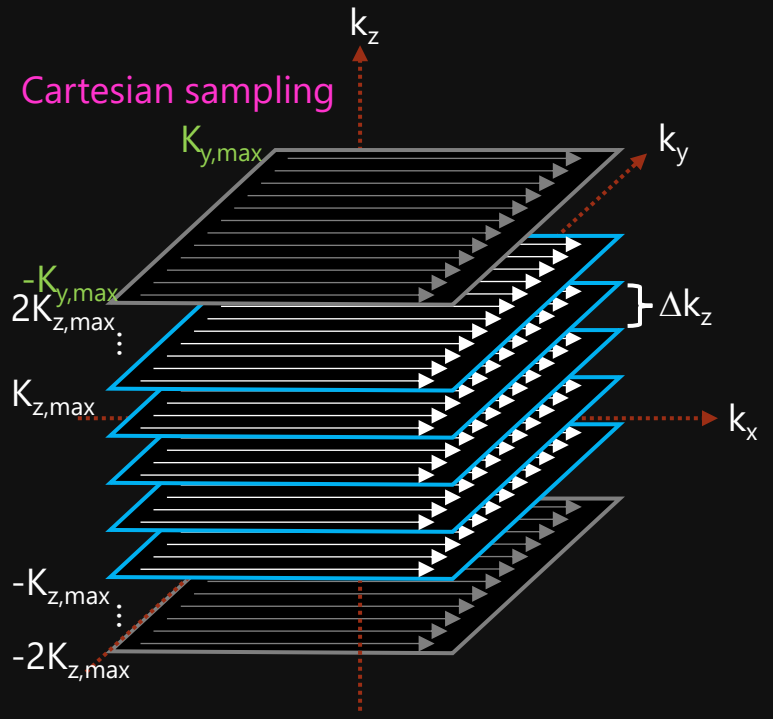
- CAIPIRINHA aims to optimize parallel imaging performance in 3D and multislice imaging
- CAIPIRINHA gives more optimal results than the normal sampling schemes, especially for very thin 3D imaging slabs or closely spaced slices in a multislice experiment.

Compressed Sensing (CS) MRI: three essential requirements and three domains involved

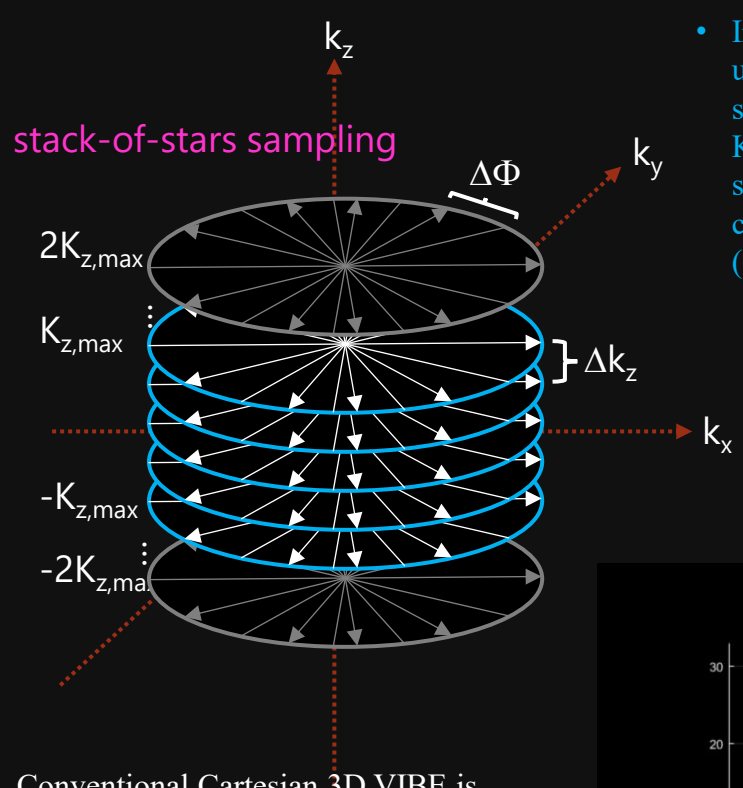


= Control parameters – thresholding, denoising, regularization, iteration number, transform sparsity, data consistency check, convergence criteria, etc.

From Cartesian to Radial to Golden-step Cartesian Acquisition



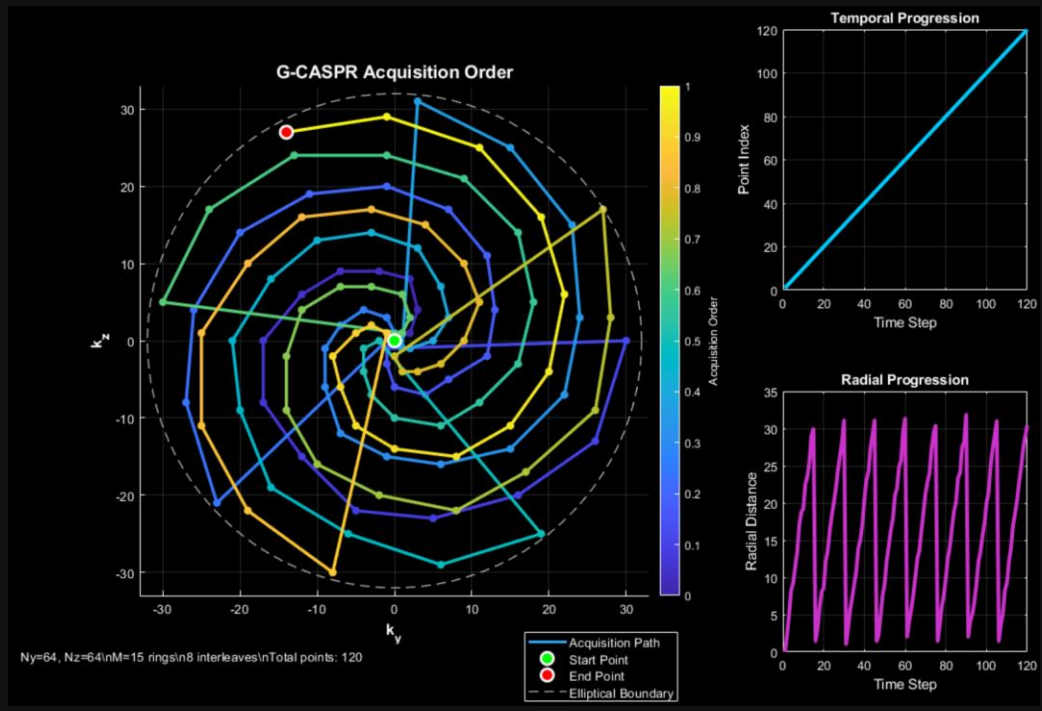
- In the conventional Cartesian row-by-row k-space sampling used in VIBE, the k-space data ranging from $-k_{z,max}$ to $k_{z,max}$ are filled with Ny × Nz data points. Along kz direction, the k-space data can be acquired in either interleaved or sequential mode. Along ky direction, sequential mode is commonly used.
- Both k-space data ranging from $-k_{z,max}$ to $-2k_{z,max}$ and k-space data ranging from $k_{z,max}$ to $2k_{z,max}$ are filled with zeros to provide small voxel size, i.e., volumetric interpolated in image space. Volume interpolation helps to minimize partial volume effect.
- The actual spatial resolution is defined by $k_{y,max}$ along the y direction and $k_{z,max}$ along the z direction.



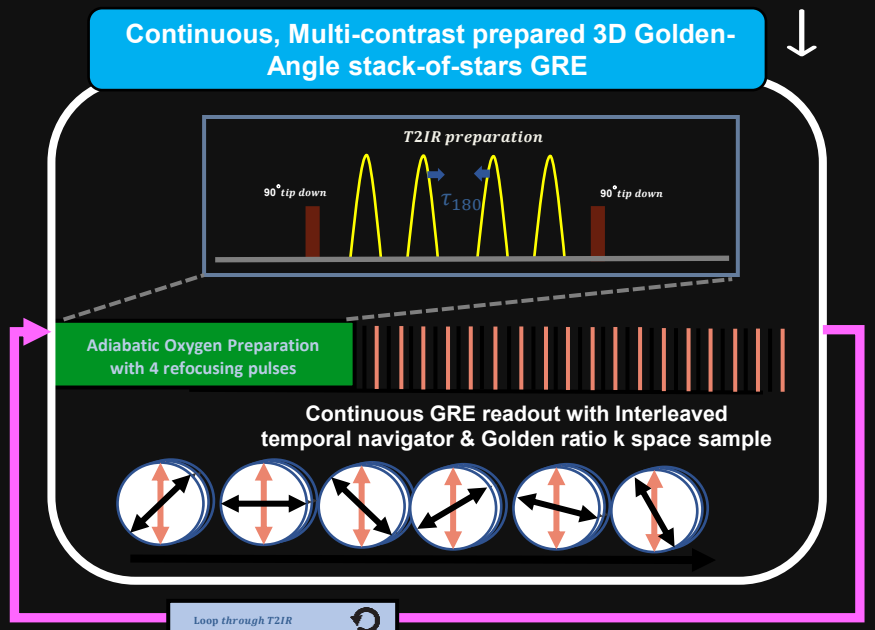
- Conventional Cartesian 3D VIBE is inherently prone to motion-induced phase distortions. Artifacts caused by respiratory motion are one of the major causes of image degradation, potentially obscuring important anatomic structures and lesions.
- Various techniques have been used to reduce respiratory artifacts when breath-hold techniques are not applicable. The two most prominent techniques are spiral trajectories and radial trajectories, which sample k-space along overlapping spokes.

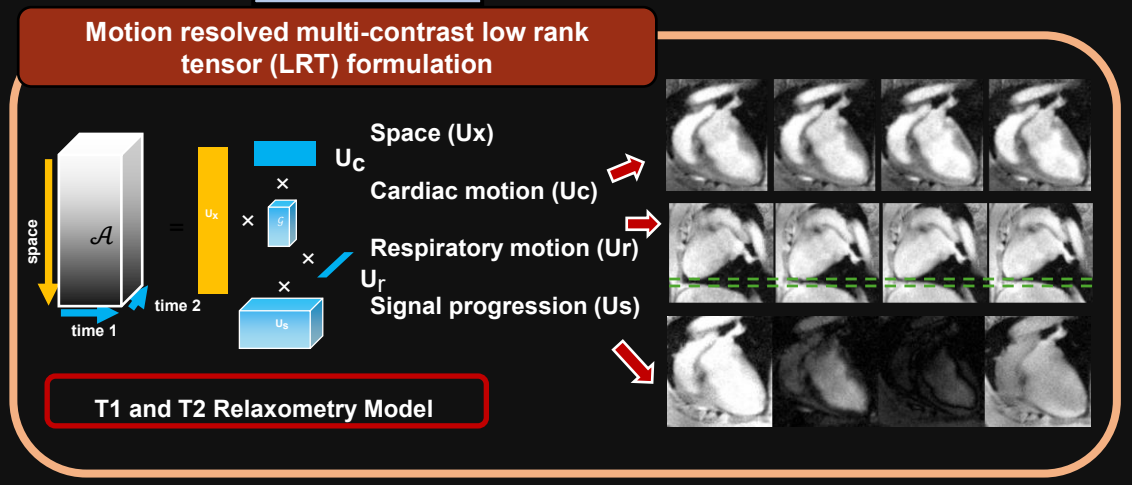
- In radial VIBE (Star VIBE), k-space sampling is implemented using a 3D "stack-of-stars" approach. In Kz (Slice) direction, standard Cartesian phase encoding is performed, while in the Kx (Read) and Ky(Phase) plane data are acquired along radial spokes that are rotated around the center, which results in cylindrical k-space coverage composed of stacked discs ("stack-of-stars").

In Golden-step Cartesian Acquisition with Spiral PRofile order (G-CASPR), image acquisition is performed using a golden-step Cartesian trajectory with spiral profile ordering. The angular step between two consecutive spiral interleaves is given by the golden ratio.



Time-resolved Whole Heart Multi-parametric mapping with model-enhanced MR Multitasking

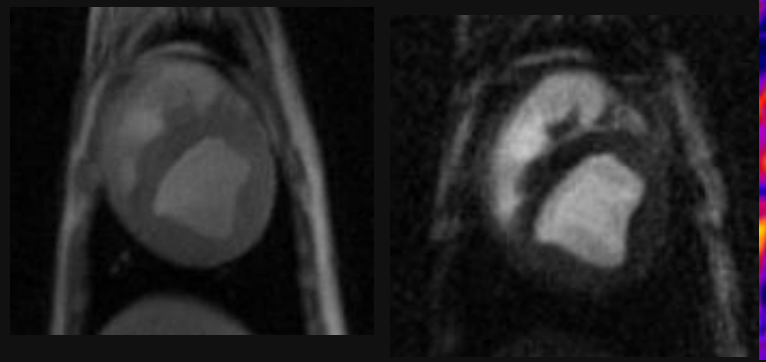




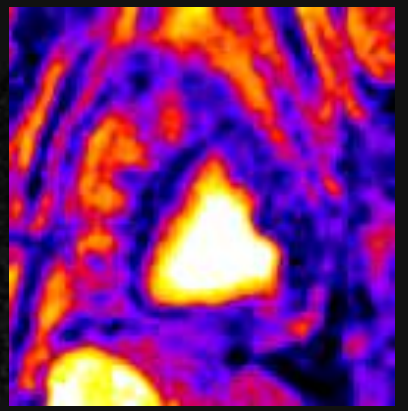
Motion resolved weighted images

Before T2prep

After T2prep



Motion resolved T2 map

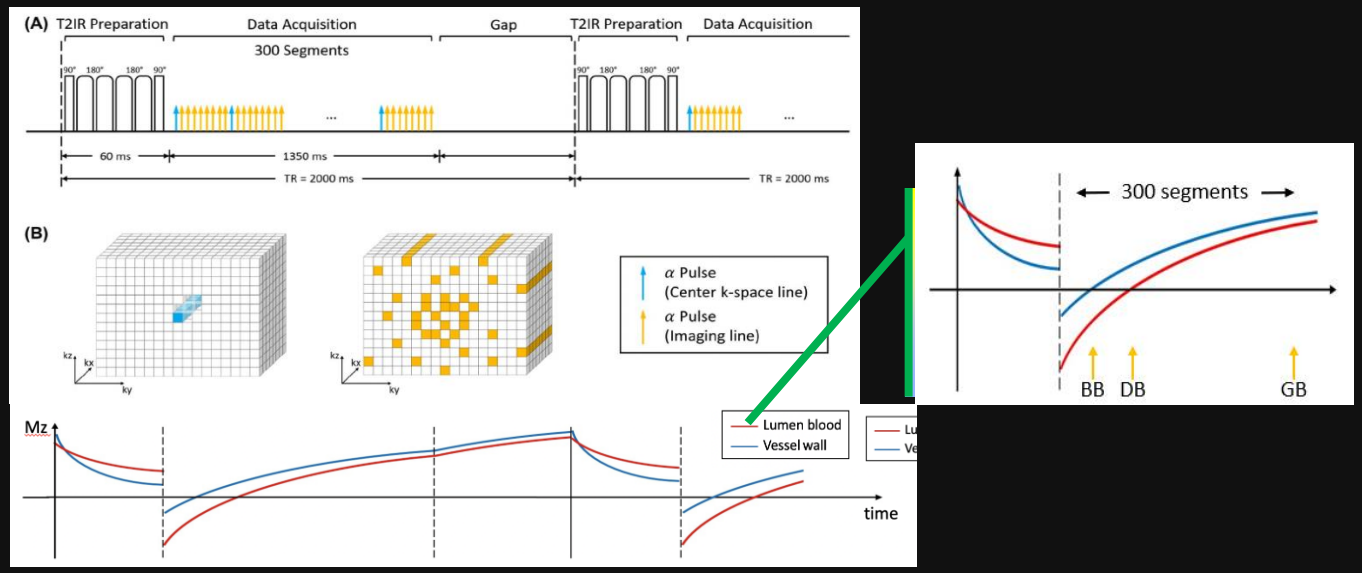


- MR Multitasking is an MRI framework that enables the simultaneous acquisition and reconstruction of multiple physiological dimensions, such as cardiac and respiratory motion, contrast dynamics, and tissue relaxation properties, within a unified, free-breathing acquisition.
- Through a continuous acquisition via different preparation modules, MR-multitasking leverages a **low-rank tensor formulation** to exploit the inherent **spatial and temporal separability** of MR signals, enabling efficient data compression and robust motion-resolved reconstruction. This facilitates efficient **multiparametric mapping** of tissue properties such as T1 and T2, while resolving motion states without needing breath-holding or external gating, making it a powerful tool for both clinical and research applications.

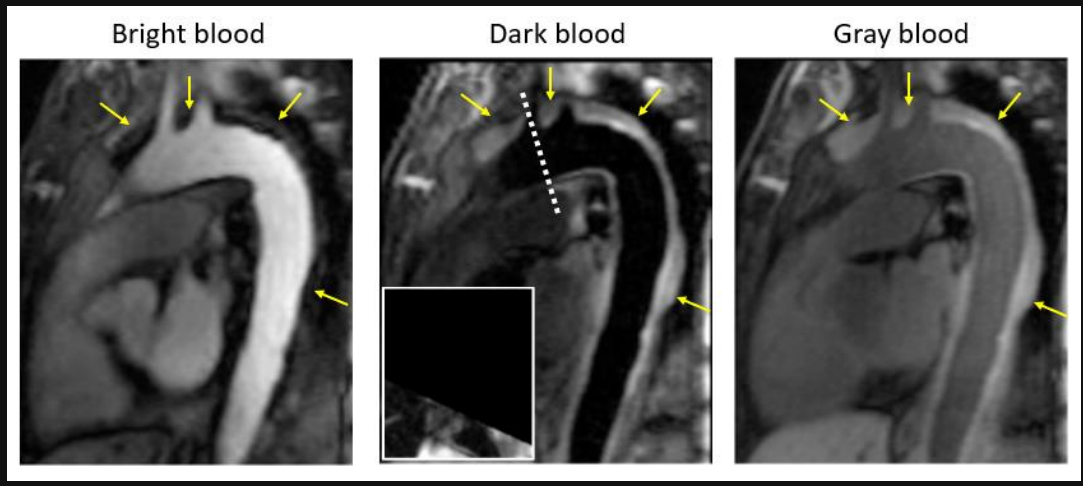
Courtesy of Dr Hsin-Jung Yang, Cedars-Sinai Medical Center

Aortic Vessel Wall Imaging

- MR-Multitasking based Multidimensional Assessment of Cardiovascular System (MACS)1,2



Aortic vasculitis



Courtesy of Dr Zhaoyang Fan, University of South California

1. Christodoulou, AG., et al. Nature Biomedical Engineering 2.4 (2018): 215. 2. Hu Z, ..., Fan Z. Magn Reson Med 2020;84:2376-2388

Acknowledgements

CAEE-60: Optimizing and Manipulating Blood-Tissue Contrast in Cardiac MRI

Zhongwei Zhang, MD, PhD: zhongweiz@wustl.edu

Jie Zheng, PhD: zhengj@wustl.edu

Scott M. Bugenhagen, MD, PhD: sbugenhagen@wustl.edu

Manish Aggarwal, MD: maggarwal@wustl.edu

Pamela K. Woodard, MD: woodardp@wustl.edu

Conflict of Interest Disclosure:

The authors declare no relevant financial relationships or conflicts of interest related to this work.

If you are interested in the topic, please feel free to reach us.

*Not all references are displayed in the EE due to the maximum slice number limit. Full list is available

Mallinckrodt Institute of Radiology, Washington University School of Medicine

Simulation and post-processing software are available: https://github.com/sljzzw