ESCOLA SUPERIOR DE TECNOLOGIA DA SAÚDE DE COIMBRA

Instituto Politécnico de Coimbra
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MRI (1.5 and 3 Tesla) sequence optimization for use in Orthopaedics

Cláudio Pereira

Mestrado em Ciências Nucleares Aplicadas na Saúde

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Dissertação

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Author's note:

Dedicated to whoever reads more than 50% of this work.

I would happily read 5 pages of this work for every page that I didn't had to write.

Published Presentations & Posters related to the following dissertation

1. **Pereira C**, Partington K. "***Reducing metal artefacts in MRI: a retrospective analysis of improved diagnostic quality and reporting confidence without the use of specialized commercial pulses/techniques***". Ed: ECR of. In. Vienna, Austria: European Society of Radiology; 2016.

2. **Pereira C**, Partington K. "***Fast scans and better images at Zero cost in Clinical MRI***". Ed: ESMRMB Congress. In. Edinburgh, Scotland: European Society of Magnetic Resonance in Medicine and Biology; 2015.

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Scientific Abstract

Background: There is currently a nonstop necessity for faster and improved imaging, in the field of clinical Magnetic Resonance Imaging (MRI). This is particularly true in Musculoskeletal (MSK) MRI, were long waiting lists are a constant problem. In addition, funds to acquire or upgrade imaging equipment have been fully or partially cut, due to the difficult financial situation that most Healthcare Systems find themselves in. The project developed aims to evaluate if and to what extent MRI Sequence Optimization can be an answer to the challenge of improving Image Quality (IQ) and decreasing scan time duration without financial investment.

Methodology: Optimized Sequences (OS) were created, for both 1.5 and 3 Tesla MRI scanners, which focused on providing the best Image Quality (IQ)/Time of Acquisition (TA) relationship. OS were developed by taking generic MRI sequences, already available in the scanners by the manufacturer, and then manipulating their MRI parameters using an iterative process in conjunction with several receiving Coils, and both non-biological and biological MRI phantoms. After the best IQ/TA relation was establish, for each sequence, they were used to create replicas of the MRI Department's standard MRI protocols. The difference in TA between the old and new MRI protocols was calculated to measure the reduction in TA obtained. IQ change was assessed through retrospective visual analysis by several blinded assessors, which had extensive experience in MSK MRI. The assessor's answers were recorded using a standard questionnaire that focused on overall IQ and also on specific technical aspects (e.g. Spatial Resolution, Contrast, Artefacts, etc.)

Results: The average reduction in TA measured was 6 minutes & 48 seconds per examination. TA reduction was more marked for protocols with a higher number of sequences. T1 was the weight type that showed a more marked TA reduction. Overall the assessors deemed that images produced from OS had either better or significantly better IQ than images produced with non-OS sequences. This trend was also present for all IQ's technical aspect assessed ($p<0.05$), with exception on Signal-to-Noise Ratio and Artefacts. T1 was considered the weight type were the most IQ improvement was observed. It was also noted that OS sequences produced higher audio noise and Specific Absorption Rate compared to non-OS, but that the safety levels were respected.

Conclusion: Sequence Optimization is indeed a useful method to improve IQ and reduce TA, without requiring any significant monetary investment. Obviously it has its limits but, if employed correctly by MRI knowledgeable technicians, it is a versatile tool that can make a significant improvement in scanner workload output and image quality.

Keywords: Magnetic Resonance Imaging; Sequence Optimization; Orthopaedics; Musculoskeletal Imaging; 1.5 Tesla; 3 Tesla; Image Quality; MRI Parameters

Resumo *(Portuguese version of Scientific Abstract)*

Contexto: Existe, actualmente, uma incessante necessidade por exames imagiológicos melhores e mais rápidos no campo da Ressonância Magnética Imagiológica (RMI) clinica. Isto é particularmente verdade em RMI músculo-esquelética (MSK), na qual longas listas de espera são um problema constante. Para além disto, fundos para adquirir ou actualizar equipamentos imagiológicos encontram-se totalmente ou parcialmente cortados, devido à difícil situação financeira na qual a maioria dos Sistemas de Saúde se encontra. Este projecto pretende avaliar se e até que ponto a Optimização de Sequências de RMI pode ser uma resposta ao desafio de melhorar a qualidade de imagem (QI) e reduzir o Tempo de Aquisição (TA) sem recurso a investimento financeiro.

Metodologia: Foram criadas Sequências Optimizadas (SO), para *scanners* de 1,5 e 3 Tesla, que se focaram em obter a melhor relação QI/TA. As SO foram desenvolvidas pegando em sequências RMI genéricas, já disponibilizadas pelos fabricantes nos *scanners*, e depois manipulando os seus parâmetros RMI através de um processo iterativo, em conjunção com várias antenas RMI receptoras e fantomas de RMI biológicos e não biológicos. Após a melhor relação QI/TA ter sido estabelecida, para cada sequência, estas foram usadas para criar réplicas dos protocolos padrão do Departamento de RMI. A diferença em TA entre os protocolos velhos e os novos foi calculada para medir a redução de TA obtida. A mudança em QI foi avaliada através análise visual retrospectiva efectuada por avaliadores cegos, os quais tinham extensa experiência em RMI MSK. As respostas dos avaliadores foram registadas através de um questionário padrão que se focou na QI geral e em aspectos técnicos específicos (exemplo: Resolução Espacial, Contraste, Artefactos, etc.)

Resultados: A redução média no TA foi de 6 minutos e 48 segundos por exame. Esta redução intensificou-se em protocolos com um maior número de sequências. T1 foi a ponderação que demonstrou maior redução do TA. No geral os avaliadores consideraram que as imagens obtidas com SO tinham melhor ou muito melhor QI que as imagens obtidas com sequências não optimizadas. Esta tendência esteve também presente para todos os aspectos técnicos de QI avaliados ($p < 0.05$), exceptuando: Relação Sinal-Ruído e Artefactos. T1 foi a ponderação na qual houve maior melhoria da QI. Foi notado que as SO produziam mais barulho e maior Taxa de Absorção Especifica que as sequências não optimizadas, embora os níveis de segurança tenham sido respeitados**.**

Conclusões: A Optimização de Sequências é um método útil para melhorar a QI e reduzir o TA, que não precisa de investimento monetário significativo. Naturalmente tem os seus limites mas, caso seja empregue correctamente por técnicos entendidos em RMI, é uma ferramenta versátil que pode fazer a diferença na qualidade de imagem e na carga de trabalho expedida pelo *scanner*.

Palavras-Chave: Imagem por Ressonância Magnética; Optimização de Sequências; Ortopedia; Imagem Músculo-esquelética; 1,5 Tesla; 3 Tesla; Qualidade de Imagem; Parâmetros de RMI

Abbreviations

2D – Two Dimensional
3D – Three Dimensional
AF – IPAT's Acceleration Factor
B_0 – Main magnetic field
BW – Bandwidth
CE – Contrast Enhanced
CNR – Contrast-to-Noise Ratio
CSF – Cerebrum-Spinal Fluid
DESS – Dual Echo Steady State
DIR – Dual Inversion Recovery
DWI – Diffusion Weight Imaging
EPI – Echo Planar Imaging
ES – Echo Spacing
Enc. Dir. – Phase Encoding Direction
FAng – Flip Angle
FAT SAT – Fat Saturation
FID – Free Induction Decayment
FISP – Fast Imaging with Steady State
FLAIR – Fluid Attenuated Inversion Recovery
FLASH – Fast Low Angle Shot
FOV – Field-of-view
FSE – Fast Spin Echo/Turbo Spin Echo
GRAPPA – Generalized Auto-calibrating Partial Parallel Acquisition
GRASE – Gradient and Spin Echo sequence
GE – Gradient Echo
HASTE – Half Fourier Single Train Echo
IQ – Image Quality
IR – Inversion Recovery
M_0 – Magnetization Vector
M_{L0} – Longitudinal Magnetization Vector
M_{T0} – Transverse Magnetization Vector
MARS – Metal Artefact Reduction Sequence
MAVRIC – Multi-Acquisition Variable Resonance Image Combination
MEDIC – Multi Echo Data Image Combination
MRCP – Magnetic Resonance Cholangiopancreatography
MRI – Magnetic Resonance Imaging
ms – Milliseconds
MSK – Musculoskeletal
NEX – Number of excitations/Averages
NOC – Nuffield Orthopaedic Centre
NMR – Nuclear Magnetic Resonance
N_{FE} – Frequency Encoding Lines/Direction
N_{PE} – Phase Encoding Lines/Direction
PD – Proton Density
PD_FS – Proton Density sequence with Fat Saturation
QA – Quality Assessment
Pixel – Picture Element
PSIF – Time-reversed FISP
RARE – Rapid Acquisition with Relaxation Enhancement
RF – Radiofrequency
RMI – Ressonância Magnética Imagiológica
rSNR – Relative Signal-to-Noise Ratio
s – Seconds
SAR – Specific Absorption Rate
SE – Spin Echo
SENSE – Sensitivity Encoding
SEMAC – Slice Encoding Metal Artefact Correction
SNR – Signal-to-Noise Ratio
SIJs – Sacroiliac Joints
SOP – Standard Operational Procedure
SPACE – Sampling Perfection with Application-optimized Contrast using different Flip Angle Evolutions
SPAIR – Spectral selective Adiabatic Inversion Recovery
SPIR – Spectral selective Inversion Recovery
STIR – Short Tau Inversion Recovery
T – Tesla
T1_FS – T1 sequence with Fat Saturation
T2_FS – T2 sequence with Fat Saturation
TA – Time of Acquisition
TF – Turbo Factor
TI – Time of Inversion (τ)
TIM – Total Image Matrix
TIR – Turbo Inversion Recovery
TMJ – Temporomandibular Joints
TSE – Turbo Spin Echo
VAT – Variable Angulation Tilt
VIBE – Volume Interpolated Breath Hold Examination
Voxel – Volume Element
vs - Versus

Section 1: Educational background

Dissertation Synopsis

Demand for faster, better and cheaper MRI imaging is at an all-time high in the Orthopaedic and Musculoskeletal healthcare environment. It is therefore important to explore any approach that reduces scan time and/or improves image quality while requiring little to no monetary investment.

In this work, the author explores how to and to what extent Sequence Optimization can be an answer to the problem presented. By applying his understanding of MRI Physics, Technology and Clinical Imaging the author analysed the clinical MRI sequences, that form the basis of the MRI protocols used in his workplace, and then adapted their parameters in order to achieve his goals.

The present written work is divided into six sections:

1. The first section relates to the educational background in which this project was developed (e.g. Synopsis, Objectives, Advisors, Authorizations, Publications, etc.);
2. The second section is dedicated to contextualizing and revising the physical principles of MRI and each of the main MRI parameters that can be manipulated by the MRI operator. It also includes some of the author's perspective/experience regarding how certain MRI parameters can and should be used in clinical imaging;
3. The third section is dedicated to detailing the methodology created and used by the author, as well as the decision-making process performed and it's justification;
4. In the fourth section the results are presented, including the methodology used to obtained and assess the results of the practical work;
5. In the fifth section the author discusses the results attained and presents the conclusions of the whole project.
6. The sixth section provides additional material, including the literary references, which supports the information and/or results obtained in the work or and help to better understand some of the content presented in previous sections. There is also present other accessory work (e.g. departmental authorizations) that the author had to perform in order to proceed with the practical component of his project. Examples of all the accessory documentation related to this work (e.g. Questionnaires, Standard Operative Procedures (SOP), etc.) are also presented.

Sinopse de Dissertação *(Portuguese version of the Synopsis)*

No actual ambiente clinico de Ortopedia e Músculo-esquelética, existe uma enorme demanda para que os exames de Ressonância Magnética Imagiológica (RMI) sejam cada vez melhores, mais rápidos e baratos. É portanto importante explorar qualquer método que consiga reduzir o tempo de aquisição e/ou melhore a qualidade de imagem, que não precise de investimento monetário significativo para ser implementado.

Neste trabalho, o autor explora como e até que ponto a Optimização de Sequências pode ser uma resposta ao problema apresentado. Aplicando a sua compreensão de Física de RMI, Tecnologia e Imagiologia Clínica, o autor analisou as sequências de RMI clínica, que formam a base dos protocolos de RMI usados no local de trabalho do autor, e depois adaptou os parâmetros, das sequências, por forma a atingir os objectivos deste projecto.

O presente trabalho escrito encontra-se dividido em seis secções:

1. A primeira secção é relativa ao contexto educacional no qual este projecto foi desenvolvido (exemplos: Sinopse, Objectivos, Supervisores, Autorizações, Publicações, etc.);
2. A segunda secção é dedicada a contextualizar e rever os princípios físicos de RMI e todos os principais parâmetros de RMI que podem ser manipulados pelo operador de RMI. Em determinados casos, também é apresentada a perspectiva/experiência do autor sobre como certos parâmetros de RMI podem e devem ser usados em imagiologia clinica;
3. A terceira secção é dedicada a expor em detalhe a metodologia criada e usada pelo autor, assim como o processo de decisão utilizado e a sua respectiva justificação;
4. Na quarta secção são apresentados os resultados, incluindo a metodologia usada para obter e analisar os resultados do trabalho prático;
5. Na quinta secção o autor discute os resultados obtidos e apresenta as conclusões do projecto inteiro;
6. A sexta secção contém material adicional, incluindo as referências literárias. Este material tem por objectivo suportar a informação e resultados obtidos para melhor compreensão do conteúdo apresentado em secções prévias. Encontram-se também expostos outros trabalhos acessórios (exemplo: Autorizações departamentais) que o autor teve de produzir de forma a poder proceder com a componente prática deste projecto. Exemplos de toda a documentação acessória, relativa a este trabalho (exemplos: Questionários, Normas de Procedimentos Operacionais, etc.) encontram-se também presentes.

Introductory note

The present paper exposes, in a formal manner, the project developed, by the author, at the Nuffield Orthopaedic Centre (NOC), Oxford University Hospitals NHS Foundation Trust. The focus of this dissertation is the optimization of Magnetic Resonance Image (MRI) sequences for application in a musculoskeletal clinical environment. Associated to this, are the underlying theoretical physics, technology and practical knowledge that support the work developed.

Additionally, an assessment of the results obtained was carried out, by way of standalone and comparative (against current standard acquired MRI images at the NOC) evaluation of the technical and diagnostic quality of the acquired MRI images. This assessment was performed by a group of qualified Radiologists and Radiographers with expertise and experience in clinical musculoskeletal MRI. Possible changes to acquisition protocols and their acquisition times were also scrutinized and, when appropriate, measured.

The author's ultimate aims with this project are:

- Present real world evidence that Sequence Optimization is a useful tool to improve both scanner workload output without requiring monetary investment;
- Improve current MRI standards at the NOC in both image quality and time efficiency;
- Obtain the title of Master in Nuclear Sciences Applied in Health, taught at the College of Health Technologies of Coimbra, under the direction of Professor Doctor Francisco Alves.

Even though this project is associated with a Portuguese institution of higher learning, it has been written solely in English do to the following reasons:

- Facilitate the dissemination of knowledge:
    - English is the Lingua Franca of the Scientific Community and any scientific work written in this language not only reaches a wider audience but also facilitates peer reviewing by the international Scientific Community.
- Ensure that both advisors could fully understand what was written by the author:
    - One of the advisors of this Dissertation is of English nationality and does not understand the Portuguese language, unlike the other advisor who is Portuguese and understands fluently both Portuguese and English;
- Maintain cohesion of language across the entire project:
    - The practical component of the project was developed in an English institution, which required that most of the accessory documentation (e.g. Questionnaires, Authorizations, etc.) to be written in English.

1 Objectives

The objectives of this project are:

1. Improve the Magnetic Resonance Imaging (MRI) images' quality, currently acquired at the Nuffield Orthopaedic Centre, Oxford University Hospitals NHS Foundation Trust;
2. Reduction of the total acquisition time of all MRI protocols;
3. Developing MRI protocols, for specific body parts and clinical situations, that might not yet exist on the NOC's MRI scanners;
4. Developing clinical sequences that respond more closely to the Radiologists' specifications and the Radiographers' needs.

2 Dissertation advisors

The project and dissertation will be guided by two advisors:

- Professor Dr Francisco Alves (College of Health Technologies of Coimbra)
- Dr Karen Partington (Nuffield Orthopaedic Centre)

3 Location

The project was developed, during the years of 2014 & 2015, in the Radiology Department's MRI service at the Nuffield Orthopaedic Centre, Oxford University Hospitals NHS Trust (Address: Windmill Road, Oxford, Oxfordshire, United Kingdom, OX3 7LD).

4 Authorizations

All the required authorizations were given prior to the start of the practical work.

All work developed adhered strictly to the OUH Trust´s policies on Safe Handling of the Equipment and Infection Control.

All the work adhered strictly to the Standard Operation Procedure (SOP) (see page 142) written specifically for this project and approved by the Clinical Governance Committee of the NOC´s Radiology Department.

Section 2: Contextualization

5 Contextualization of MRI in today's clinical setting

In today's Medical setting, MRI has established itself as an essential, or at least highly valuable, asset to have in the majority of clinical cases. From Research to Clinical work, MRI plays a major role in pretty much all areas of medicine. Used to screen for early signs of pathology, diagnostics, treatment evaluation and disease follow-up, it quickly becomes hard to find an area of medicine where MRI isn't a useful tool.

No greater proof of this exists than the fact that the term "MRI" is, nowadays, quite familiar to the average population and its association with the idea of being an "all powerful wonderful healing machine". This idea is clearly misguided, yet it is also a testament of the important role that MRI has in today's medical environment.

For those who never had the opportunity to study this field of knowledge, MRI presents itself as an intangible bizarre conundrum. For those fortunate (or not) to dwell in it, MRI becomes a highly complex multileveled structural puzzle that gets more understandable and beautiful with each piece that one can connect in its rightful place, while at the same time presenting more intricate problems to be solved.

Showing an astounding technological development rate, the end of MRI's potential is still hard to see on the horizon and there is no greater beneficiary of all this improvement than clinical medicine.

5.1 The interest and advantages of sequence optimization

With the current increased demand for MRI examinations, even considering the fast growth of the MRI market, it is plain that there is great need for an increase in workflow and output within MRI departments. This need is obvious to anyone involved in clinical MRI. A need, so significant, that is, today, one of the flagships that every MRI manufacturer tries to achieve or, at least, pretends to provide.

Radiographers, being on the frontline of MRI clinical practice, deal with this problem every single day and will certainly have a complex answer to the question: How can we improve MRI workflow without increasing costs?

Personally, there are several things that can be done in multiple areas of MRI, but the most important is definitely sequence optimization. By reducing scanning time duration, while keeping image quality constant, we could try to give a "simple" solution to the problem of demand exceeding the offer.

However, at the same time, a need for increase in image quality presents itself, giving the problem additional complexity. It's when trying to conciliate faster scanning with better imaging that the need for sequence optimization (and so the relevance of this work) emerges as the only reliable, long term and low cost solution.

6 MRI's Physical Principles

It's important to bear in mind that when speaking about MRI physics we are, in fact, talking about Nuclear Magnetic Resonance, which derives from that unfathomable box of physical weirdness that is Quantum Mechanics.

The physical principles of Nuclear Magnetic Resonance (NMR) have been known since 1947, due to the works of Felix Bloch and Edward Mills Purcell [10]. Even its application in medicine has been advocated for several decades, tracing back to J. R. Singer, in 1959, that suggested its use as a non-invasive method for measuring in vivo blood flow [11].

Clinical images based on NMR only appeared in 1977 through the experiments done by Paul Laterbur, Peter Mansfield and Raymond Damadian. The first actual MRI was obtained by Paul Laterbur using a NMR spectroscope and consisted of a very small crustacean in a tube filled with heavy water. [10]

However, more important than this are the actual physics at play.

6.1 Nucleons, Spin and Stable Energy States

It all begins with nucleons and their behaviour under certain conditions in atoms that possess an incomplete filling its nucleus's layers. Nucleons (Protons or Neutrons) are particles that constitute the atomic nucleus and possess an intrinsic property called: magnetic moment (Protons = ½ and Neutrons = 0). Normally atoms fill their nuclear layers in pairs of two nucleons, with the particles that compose each pair being aligned antiparallel to each other. This nullifies the magnetic moment of each nucleon, creating atoms that, by themselves, show no magnetic properties. On the rare occasions when an atom has an odd mass number (like in a hydrogen atom) the magnetic moment differs from zero giving the atom its own magnetic moment or Spin, as it's often called. [7, 10-12]

When the Spin of an atom differs from zero, the atom behaves as a tiny eternally wobbling (this motion is known as precession) magnet that, when exposed to a strong external magnetic field (B_0), is forced to align itself with the strong field in, either, a parallel or antiparallel direction.[7, 10, 11] The distinction between the two alignments is only energetic, being the parallel one the lower energy state. This allows for the two types of alignment to be treated as energy levels instead of physical orientations.[7, 10, 11, 13] Both of which are stable energy states.[11] Atoms can then swap their energy state (alignment) if excited with energy equal or higher than the difference between the two energetic states. [7, 10, 11, 13]

The statistical distribution of protons, between the two energy states, is a random probabilistic process influenced by the proton´s internal energy, B_o, the thermic energy present in the system. In a human body (37ºC) at B_o equal to 1,5T, the difference in the number of protons in both energy states is only 4 (in favour of the low energy state) per million of atoms in each energy state.[11] This difference in the distribution of the protons increases in higher field strengths and [7, 11, 13]

The energy difference between the two states increases with the increase of B_0, which is the reason behind the increase in intrinsic signal observed at higher magnetic fields.[5, 7, 11, 12]

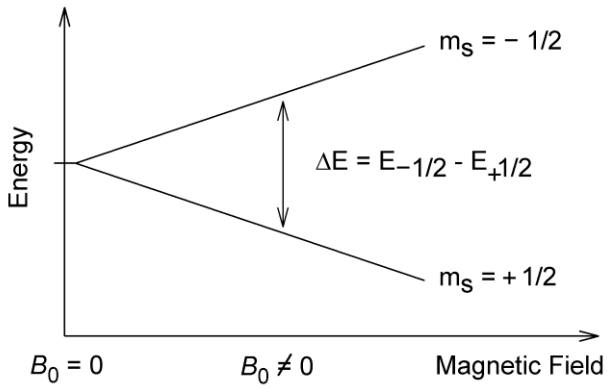


Figure 1: Relation between B0 and energy difference between the two energy states. Extracted from [14]

6.2 Resonance Principles and Precession Frequency

The nucleus of every atom precesses at a specific frequency (called the Larmor Frequency (Equation 1)) and because the non-neutral Spin of some atoms causes them to interact with the external magnetic field, the frequency at which those atoms precess becomes dependent of the intensity of the field.[7, 10, 11, 13]

This precession isn´t a physical motion of the atom/proton, but a variation in the direction of the atom/proton Spin's vector.

$$Larmor\ Frequency = B_0 * \gamma/2\pi$$

Equation 1: Mathematical formula for the Larmor Frequency of an atom. B0 represents the strength of the external magnetic field and γ the gyromagnetic ratio of the atom.

This is amazing because it creates a system where we can interact with large groups of atoms, enabling us to obtain a reliable feedback from them. Interacting with only one proton would give a feedback signal too feeble to work with.

What is the meaning of interact in this context? Well, since atoms precess at a specific frequency, they are very responsive to any wave/stimulus that has the same frequency as them. This phenomena is defined in physics as the Resonance, hence the meaning of the R in MRI and NMR.

Additionally, besides exciting protons, some stimulus (Radiofrequency (RF) pulses) can also force the nuclei to precess in-phase with each other.[7, 11, 13]

This doesn't imply that they are insensitive to others frequencies. It means that you get a much stronger reaction from a certain group of atoms the closer you are to their Larmor Frequency.[7, 11, 13] It works like a swing. A swing moves around in an ordinary motion with a specific rhythm. If someone tries to push the swing, two things may happen depending on how we push: if someone pushes the swing at a time and rhythm different from the swing's rhythm it will destroy the motion of the swing and destabilize the system; if someone pushes the swing at the right time and rhythm then not only the swing's motion will become more stable, it will also swing more.

This phenomenon doesn't apply only to atoms and waves.

From a straight physics point-of-view it's a situation where one can transfer energy to a system/object making it very excited, but it can only done with RF pulses that have a specific frequency (sending a RF pulse with a frequency equal or very similar to the proton's Larmor frequency).

6.3 Relaxation: Recover, Decay and Dephasing

Exciting protons is quite fun but it's also meaningless, unless you get a reply that you can work with. That is exactly when one of Physics´ most basic principles comes in. All physic systems tend toward stability and this is normally found at lower energy levels.[7, 10-13] In NMR's case, excited protons have to release the energy that we transferred to them. In MRI this process of releasing energy to the environment is called Relaxation, meaning that an atom or group of atoms after being excited will precess more "steadily" while, at the same time, "slowly" returning to their natural precession state (one of the two stable energy states). [7, 10-13] While this occurs the nuclei release energy into the rest of the system. It's from this process of relaxation that we can "capture" our reply/feedback. By having an antenna (coil) near the vicinity of the excited atoms, the relaxation process will induce (Figure 2) an electromagnetic signal in the coil.[7, 10, 11, 13]

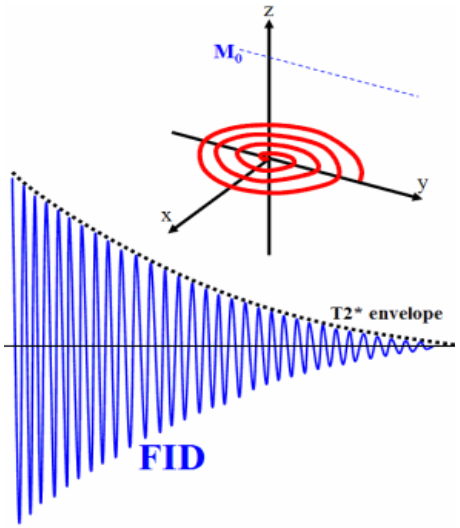


Figure 2: Induced signal from an atom´s FID (only Transversal Magnetic Vector considerer in this example). Extrated from [7]

The signal induced can be sampled, registered and processed as a reply from the atoms.[2, 5, 7, 8, 11, 13, 15-17] This signal reflects the way the atoms return to a more stable energy level, and this dependent on the conditions/environment that the protons find themselves.[5, 7, 10-13]

The signal's wave shape is dimensionally quite complex due to the variable direction, angulation and size of the Spin's vector (Figure 3). To simplify everything, the Spin's vector is split into two simpler vectors that have constant direction: the Longitudinal Magnetization Vector (ML_0) and the Transverse Magnetization Vector (MT_0). By separating the longitudinal component from the transversal component of the Spin's vector, it becomes much simpler to process the signal and also allows for acquisition of multiple types of contrasts. By treating the ML and MT separately, instead of dealing directly with the Spin's vector as a whole, we can control more precisely what type of contrast we desire. It's from this that we get what we call: Image's Weight (see Chapter 0)

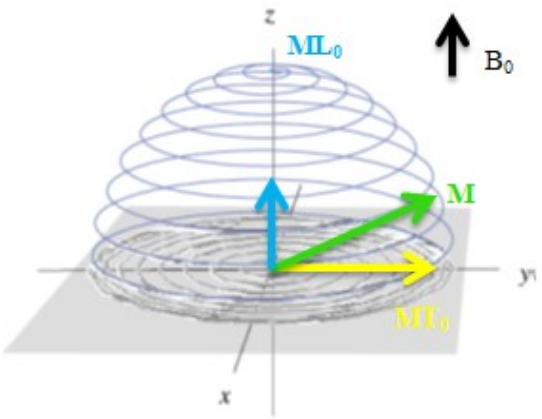


Figure 3: Magnetization Vector's (M0) real trajectory (Green arrow) during relaxation. The direction tends towards Z (same as B_0). The vector can be divided into two other (simpler) vectors that represent the longitudinal (Blue arrow) and transversal (Yellow arrow) components of the real vector. Adapted from [11]

The process of Relaxation (Figure 5) can be divided into three components:

1. Recover: Reflects the way how the Net Magnetization (sum of the ML_0 of all excited protons) recovers over time. It contains the information relating to the Spin-Lattice relaxation time (T1) of the tissues. [11, 13] It's maximal when the protons regain their natural stable state.
2. Decay: Main component of the Spin-Spin relaxation time (T2).[11, 13] Exposes the fashion how the MT_0 of the protons diminish over time during the relaxation period. It describes the Free Induction Decayment (FID) (Figure 2) which defines T2*.[7]
3. Dephasing: Reflects that nuclei tend toward non-organized (non in-phase) precession (Figure 4). It's the second contributor (in an invert fashion) to T2 relaxation. When a group of nuclei precess in phase with each other, constructive interaction between them occurs making the nuclei less susceptible to B_0 inhomogeneities and other factors that speed up the MT_0 decay. As a result the effective Spin-Spin relaxation period is extended, becoming true T2 decayment, instead of FID (T2*).

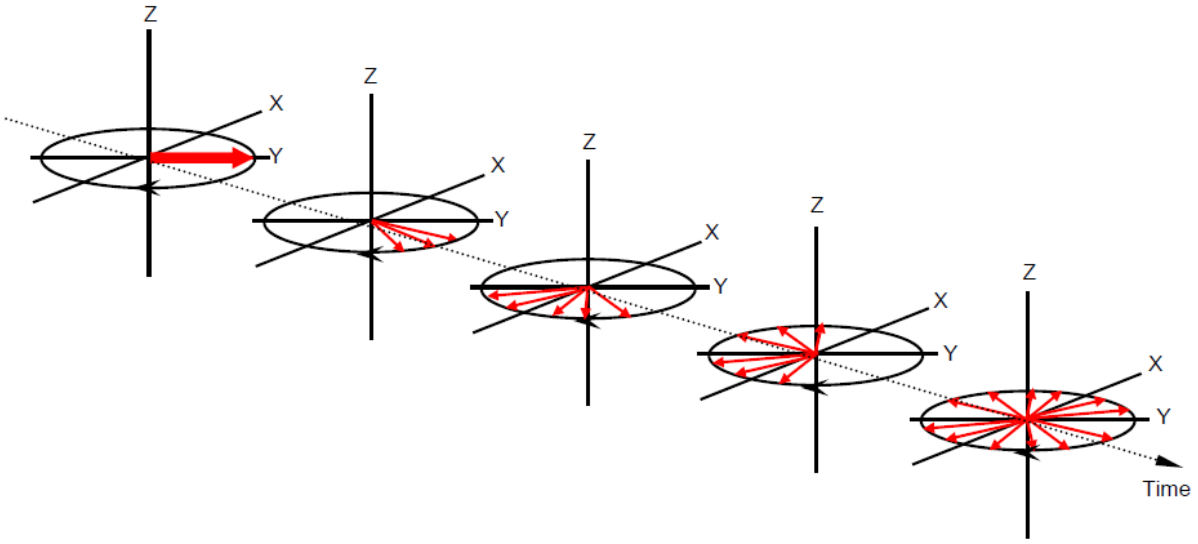


Figure 4: Atoms' spins dephasing between each other over time. Modified from[18]

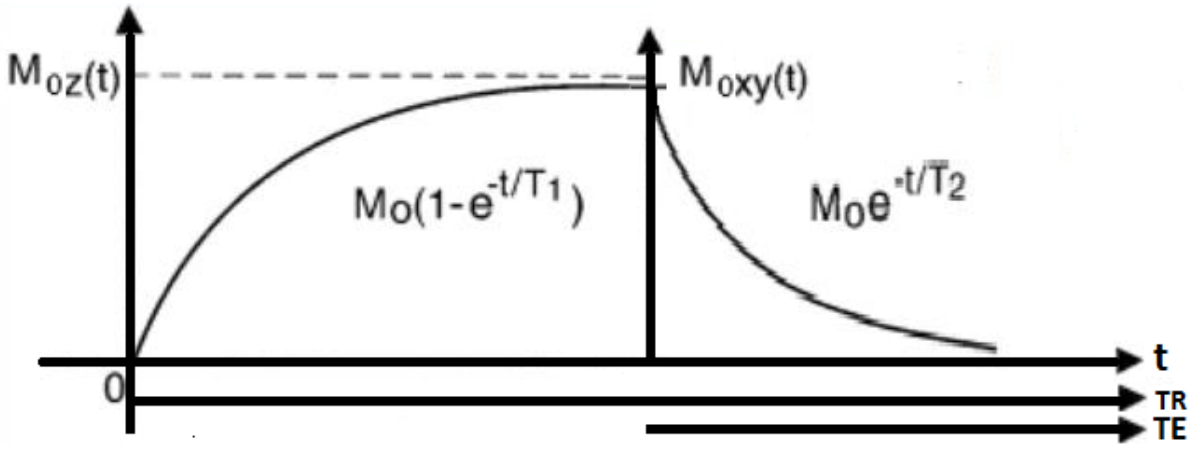


Figure 5: Relaxation curves for both ML and MT components of the proton's Spin Vector.

Spin-Spin interaction occurs because nuclei are in fact particles with electrical charge.[7, 11, 13] When nuclei precess they became effectively moving charges and so each proton possesses its own magnetic field. The field is extremely small and weak, but it can still affect other electrically charged particles in its reach (protons, electrons, etc.).[11] The type of interaction is dependent on the direction that the nuclei are precessing and the alignment of their phases.[7, 11, 13] It will be constructive only if the two (or more) nuclei have the same precessing direction or are in-phase with each other. It will be destructive if protons have opposed precessing direction or phases).

Both types of interaction can be advantageous depending on the desired aim of the proton manipulation.

Groups of protons with the same precessing direction can be obtained by excitation through RF pulses (because they possess rephasing properties).[11, 13] The opposite can be obtained by application of the spoiling/dephasing gradient.[11, 13]

Spin-Spin interaction, alongside phase alignment, is fundamental in MRI sequences for generating echoes, destroying magnetization/signal (e.g. spoiling), phase-based contrast, etc.

6.4 Why Hydrogen-based MRI

Hydrogen is not only a very simple element, with a low Larmor Frequency, but is also an extremely common element in the universe and in the molecules that are the building blocks of our body, making it the best element to use in MRI.[7, 11]

This doesn't mean that MRI only makes use of hydrogen. It would be a waste of potential to ignore all the other useable elements, but clinically speaking due to Signal-to-Noise (SNR) related limitations, nowadays, hydrogen is king and it's very likely to stay that way for a long time.[11]

Hydrogen is characterized by a Larmor frequency of 42.52 MHz per Tesla [5, 11] (or 63.855 MHz at 1,5T), requiring very low energy to excite it. It is also very common in our bodies, meaning that good SNR is obtainable, and because its concentration and the way it binds with other atoms differs depending on the biological tissue where the atom is present, we get in one go, easy signal acquisition, high SNR and excellent contrast between body tissues.

In a weird comparison, Hydrogen in MRI it's like 18-FDG in PET, great in everything except specificity.

On a side note, good specificity is why non-hydrogen MRI is also interesting, since specificity is of great interest for diagnostic purposes and therapy follow-up.

6.5 Technical and Technological Principles

On a technical perspective there are four main aspects in MRI to obtain an image:

1. Aligning protons using high intensity magnetic field (meaning putting a patient on the MRI machine, making him/her into a magnet);
2. Using coils to send RF pulses to the body and coils to receive/sample the reply from the body (exciting protons and capturing the signal that they release);
3. Encoding the signal in space using magnetic gradients (defining slices and filling the K-Space for each slice);
4. Reconstruct mathematically the information in K-Space into Real Space images.

6.5.1 RF pulses

Radiofrequency pulses are electromagnetic in origin and are used in MRI to manipulate the protons in both energy states. They can be used to excite (partially or fully), invert, saturate, rephrase, or modulate the signal intensity of the protons.[4, 7, 11, 13, 19, 20] Their shape can be a simple or complex function, with various degrees of amplitude and frequency.[19, 20]

RF pulses interact with protons through the principles of quantum mechanical radiation-matter interaction, where energy from an electromagnetic wave can be absorbed or emitted by quantum particles that precess at a similar frequency than the wave itself. [4, 7, 11, 13, 19, 20]

Refocusing of protons always occurs during application of RF pulses.

Usually in MRI, RF pulses are defined by the degree of angular tilt they force upon the proton´s spin vector (e.g. 90º pulse).[7, 11, 13] The spin vector´s tilt (in degrees) directly correlates with the level of the protons' excitation, inferring that pulses of certain degrees can actually represent different proton states:

- Excitation pulse - A 90º pulse applied when the protons are relaxed, producing full excitation of protons.[5, 7, 11, 13, 19] If lower than 90º then it's just partial excitation (assuming the protons are relaxed at the moment when the pulse is applied);
- Inversion pulse - A 180º RF pulse applied when the protons are relaxed. The energy states where the protons are initially located are swapped.[5, 7, 11, 13, 19] (the system is unstable because there are more protons at the higher energy stable state than in the lower energy state);
- Refocusing pulse – A 180' RF pulse applied during the relaxation period. The direction towards the protons are dephasing is maintained.[5, 7, 11, 13, 19] The protons are swapped to a mirror position (in the XY axis) of their original location (when the pulse was applied). This is known as the Hahn condition for Spin Echo creation.[21]

6.5.2 Signal acquisition, Gradients and image encoding

Signal acquisition is done by digitally sampling the analogous signal wave induced in the receiving coils.[4, 5, 7, 11, 13, 15, 16, 19] Sampling provides the intensity value of the signal and is the basis of the measured/real SNR. The location of the nuclei, from where the signal is coming from, is hard to spatially locate through sampling alone, as explained by the Uncertainty Principle that rules over all quantum systems.

To obtain the precise location of the signal's origin, strong electromagnetic gradients must be employed. Electromagnetic gradients are created by physical coils inside the scanner and can play many rolls in MRI, but their main usefulness is to encode the signal in space. [4, 5, 7, 11, 13, 15, 16, 19]

Three gradients, aligned orthogonally to each other, are employed to generate bi-dimensional and "tri-dimensional" images. They behave as electromagnetic slopes that slightly disturb B_0 in space (in a known fashion).[4, 5, 11, 13] The known B_0 disturbance causes protons, which are in different levels of the gradient, to precess at slightly different frequencies/phases.

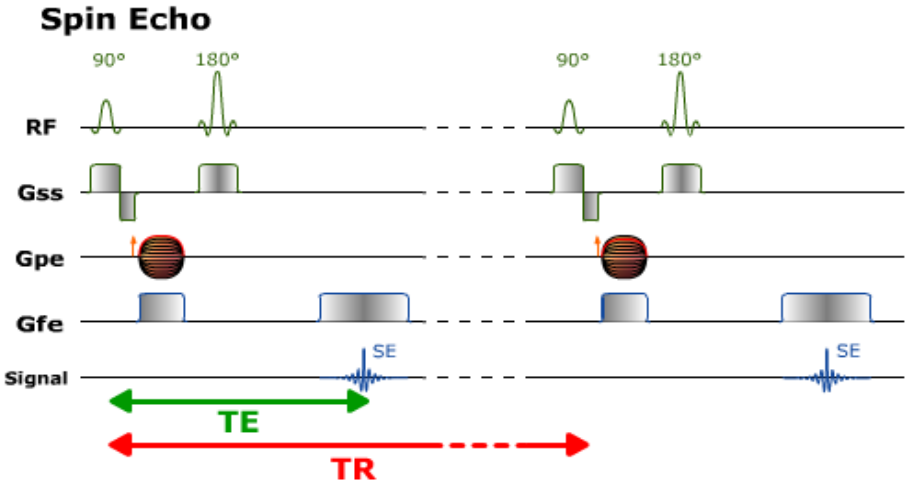


Figure 6: Spin Echo sequence diagram. Radiofrequency pulse (RF); Slice Selection Gradient (Gss); Phase Encoding Gradient (Gpe); Frequency Encoding Gradient (Gfe). Extracted from [13]

If the gradient's shape resembles a single step/bump, at the time a RF pulse is applied, then selective excitation of a region of atoms becomes possible.[3, 11, 13] Unfortunately, applying a non-modulated sinusoidal RF pulse will produce a poor defined slice. This can result in partial volume artefact and in cross-talk artefacts (see subchapter: 14.2.1.6) (when multiple slices as simultaneously selected and close to each other (see subchapter: 9.2.6)). Well defined slice selection can be achieved by using modulating the RF Pulse with a SINC Pulse (or a truncated approximation of one) (Figure 7).[3]

The combination of the gradients generates a virtual tri-dimensional grid system that permeates the scanned object. This allows the system to discriminate the origin of the signal based on the gradient coordinates of the voxels that the 3D grid is comprised of.[3-5, 8, 10, 11, 13, 18]

Classically the XYZ axes represent frequency, phase and slice respectively, but the gradients flexibility makes them more or less independent of the real orientation of the system. In 3D scanning there is no slice, instead the excitation is performed per Slab (which represents a volume approximately equal to the entire Field-of-View (FOV)) and Z behaves as a second phase axis.[5, 11, 13]

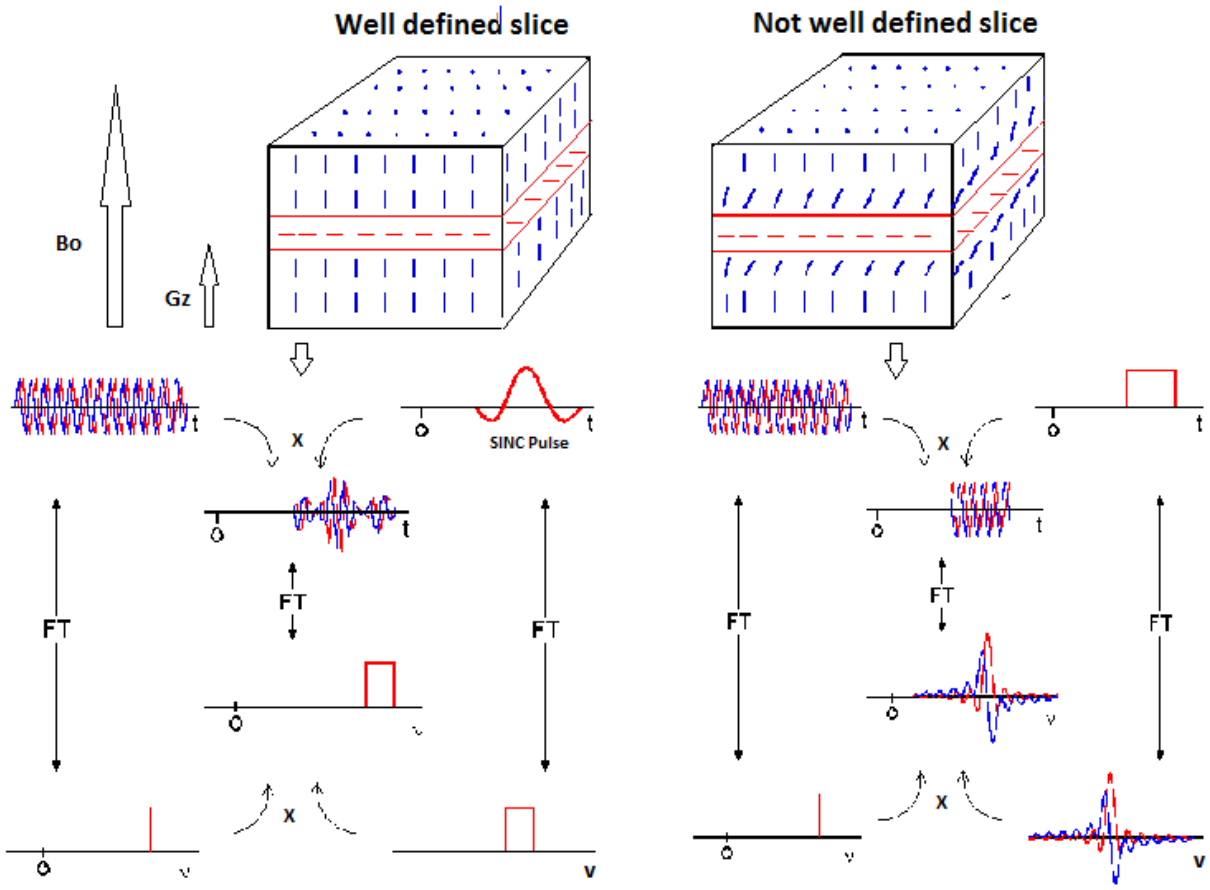


Figure 7: Slice selection. Extracted from [3]

6.5.3 Why use Echoes instead of FID?

When excitation occurs, decayment starts almost as soon as the RF excitation pulse ceases. [7, 13] This decayment is not an echo, but FID. It occurs spontaneously and it's hard to control or manipulate. FID has also a very short interval of time, after the RF pulse is shut down, making it hard to obtain signal in the receiving coils that is not distorted by the RF excitation pulse (through wave constructive-destructive interference) and to sample the signal at the peak of its amplitude/intensity.

Echoes on the other hand are quite distant (time wise) from the excitation pulse, preventing any interaction between the two (on the receiving coil). Additionally echoes are artificially generated, possess a time length approximately twice as long as FID and the signal's maximal intensity is achieved in the middle of the echo wave (FID's signal peak in always at the beginning of the wave). This makes an echo more suitable for controlling the image's weight and signal sampling.

The drawbacks of using echoes to generate images are a significant increase in the minimal TR and reduced SNR. Minimal TR is increased because echoes are longer than the FID and can only be generated sometime after the FID's end. SNR is reduced because the amplitude of the echo is much smaller than the FID's amplitude. This amplitude reduction increases with higher number of echoes created within the same TR, due to imperfections in the refocusing pulses and atom saturation.[11]

6.5.4 MRI coils.

Apart from the sequences themselves, MRI coils are the single most important aspect to consider, to guaranty minimal image quality.

From the coils used, to the way they are positioned on the patient and the maintenance they receive, MRI coils can "make or break" an examination. Selecting the wrong coil, placing the coil or its cable in the wrong place or way, and even previous, unrepaired, physical/electrical damage to the coil will prevent the acquisition of good images in acceptable time.

Being such a critical factor, it's then essential to comprehend how MRI coils intervene in the process of signal acquisition and what type of coils exist. Detailed information about each coil, how to use and test them is normally provided by the manufacturer in the form of a manual.[5, 16]

MRI coils are used to transmit or receive signal. Transmitting coils send the RF pulses to excite/invert/saturate/refocus the protons. Receiving coils "capture" the signal released by the protons when they are relaxing.[5, 11, 16, 19] This "captured signal" is in fact electrical current that is induced, by the relaxation RF wave, within the conductive material that the receiving coil is made of.

All MRI scanners possess at least one transmission/receiving coil incorporated within them. This coil is normally the one used to excite the protons in the majority of the examinations. The name of this coil varies according to the scanner's maker, but is usually defined as Body/Q/Transmit coil.[5, 16] Receiver coils are the ones secured on to the patient before the beginning of the scan.[5, 16]

Most coils only transmit or receive signal, but some are capable of performing both tasks and as a rule of thumb they tend to be best ones, due to higher SNR, more efficient excitation and lower SNR. [5, 16]

Speaking in a very generic way, receiving coils can be divided in two categories: Dedicated Coils and General Purpose Coils. Obviously this isn't a technical division, but in clinical practice it is a simple and functional way to distinguish coils, rather than having to separate coils as surface coils, quadrature coils, etc.

Dedicated Coils are normally "small" in size and shaped in concordance with the body part that they are meant to be used on. They usually possess more channels than other coils and sometimes are both emitter and receiver.

General Purpose Coils tend to vary significantly in size, have a rectangular or circular shape, be somewhat flexible and be able to be coupled with other coils.

From a technical point-of-view, Dedicated Coils are better than General Purpose Coils, since they provide better SNR per cubed centimetre and less sensitive to external interferences. This is the result of a combination of factors, chief among them are the higher number of channels, emitter/receiver capabilities and better tuning/shimming.

In Musculoskeletal (MSK) examinations the relative position to the body, shape and size of the scanned structure varies greatly. It is then impossible to rely only on Dedicated Coils. Additionally, limitations in patient positioning, due to some patient's inability to cope with certain positions for a long time, vastly limit the number of situations where Dedicated Coils can be employed.

As a practical example let's consider an ankle examination to query the extent of an active infection:

Usually, ankles are scanned using a Dedicated Foot/Ankle Coil with the patient in feet first dorsal decubitus. This is a very stable and comfortable position for most patients and the size of the coil enable it to be used for both small and large ankles. In some situations though, the patient may be experiencing too much pain on the heel to withstand having it supporting the weight of the foot. In this situation, and assuming that the patient has both some range of motion on his ankle and a small to medium-sized ankle, the examination can be performed with the patient in ventral decubitus and using a dedicated Knee Coil. This change in positioning and coil allows for an examination where the patient is cooperative and a high SNR coil is employed, making the acquisition of good quality images feasible.

It can also be the case that the same patient has pain in the entire ankle, reaching the midfoot and the distal third of the leg. In this case an assumption needs to be made by the radiographer that the infection is actually extending beyond the ankle and so a larger FOV must be used.

Dedicated Coils have a limited maximal effective FOV, meaning that they have high SNR within a specific volume but that SNR drops quickly beyond that volume. This characteristic make the coil unsuited for scanning large body parts, since image quality will vary greatly between images in the same plane and within the FOV. In this case using General Purpose Coil(s), that cover both the entire foot and the lower third/half of the leg, is the best option since it will guarantee full coverage of the infected tissue and deliver a more homogeneous image quality (even though the image quality is lower that would be achievable with a Dedicated Coil).

The same choice should be made if the degree of swelling is too severe for the foot to properly fit in the dedicated coil.

Ultimately is becomes a case-to-case scenario and it may well be extremely useful to have two or more protocols, for the same body part, that are optimized for different types of coils, positioning and patient size.

Generically, for any given examination use the coil(s) that:

1. Covers the entire area of interest;
2. The patient feels more comfortable with;
3. Allows for the scanned body part to be closer to centre of the bore;
4. Has the highest number of channels;
5. Is IPAT compatible;
6. Require associating the lowest number of coils possible (one shouldn't use more than 2 unless you are covering a very extensive area (example: Neuro-axis requires Head, Neck and Spine coils));
7. You are more familiar with.

6.5.5 Sampling signal and the Nyquist Theorem

Under the rug of signal acquisition lays the tedious process of analogue-digital signal/data conversion/sampling. It is here that the Nyquist Theorem makes its crucial appearance.

When converting an analogue signal into digital data, it is essential to make sure that the sampled signal is the same as the original analogue signal. This can be achieved by making sure that the rate of signal sampling is high enough to fulfil the Nyquist Theorem.[8, 11, 15, 22]

$$F_s \geq 2 * F_{max}$$

Equation 2: Nyquist formula: Fs (sampling frequency); Fmax (highest frequency in the original analogue signal.

Failure to comply with this theorem will lead to Aliasing-related artefacts (see subchapter 14.2.1.1).[8, 11, 15, 22]

In MRI the sampling rate can be manipulated through the receiving Bandwidth (rBW), with increased Bandwidth (BW) leading to improved sampling rate.[15]

7 Basic image Weights: T1, T2, PD and T2*

The word "Contrast" can have different meanings when analysing an image. Usually, Contrast refers to the ability to visually discriminate two adjacent Picture Elements (Pixels) based solely in the difference between their intensity values. It is a qualitative characteristic of the image. In reality Contrast is a complex concept that can include the pixel's size (image's resolution) and/or the number of possible values that the pixel can have (the image's grey scale).

In MRI, however, contrast can also refers to what component of the relaxation process dominates the signal intensity of the different tissues. To avoid confusion this is referred as the image's Weight (Figure 10)

As exposed before, tissues in MRI can have very different intensities depending on what time, in the relaxation process, the signal is sampled. Additionally different tissues possess different relaxation curves and are affected differently by the type of echo created and pre-pulses applied. It is therefore essential to contextualize the pixels' intensities, seen on an image, with the dominant relaxation time in order to be able to correlate reliably the intensity seen, on a given pixel, with the respective tissue it represents. Without this contextualization it would be impossible to be sure what tissue the pixel represents because the possible intensity values for a given tissue can vary wildly. For example: water can appear black, dark grey, light grey, bright white or very bright white depending on how much contribution there is from MT_0 to the overall weight of the image.

To better comprehend how each tissue's signal decays overtime, two concepts were created: time of relaxation related to ML (T1) and time of relaxation related to MT (T2) (Figure 5 & Figure 8).

The T1 and T2 of a specific tissue (

Table 1) refers to the time (in milliseconds (ms)) it takes for the hydrogen nuclei within a tissue to achieve a specific percentage (63% for T1 and 37% for T2) of its original Magnetization

Vectors (MT for T1 and ML for T2), after a 90º RF pulse is applied (Figure 5 & Figure 8). Tissues with long T1 will have low intensity in T1 images (e.g. Water) and tissues with short T1 will have high intensity of T1 images. In opposition, tissues with long T2 will have high intensity values on T2 images and tissues with short T2 will have low intensity values.[5, 7, 10, 11, 13, 18, 23]

The opposition exposed results from the angular direction that both vectors tend to go over time. ML goes from a null value towards a positive one, while MT goes from positive towards a close to null value (Figure 5 & Figure 8).

As mentioned before, by carefully selecting the point in time when signal is sampled, it's possible to obtain a T1, PD or T2 Weight (Figure 9 & Figure 10).[5, 7, 10, 11, 13, 18, 23, 24]

Table 1: T1, T2 and Rho of different tissues at multiple field strengths. Extracted from [5]

	on 1.0T systems			on 1.5T systems			on 3.0T systems		
Tissue	T1 [ms]	T2 [ms]	Rho [%]	T1 [ms]	T2 [ms]	Rho [%]	T1 [ms]	T2 [ms]	Rho [%]
Grey matter	772	100	85	920	100	85	1200	100	85
White matter	650	90	70	780	90	70	1010	90	70
CSF	2100	160	100	2400	160	100	3120	160	100
Skeletal muscle	710	45	70	870	45	70	1161	45	70
Vertebral marrow	375	60	40	400	60	40	520	60	40
Bone marrow	375	60	40	400	60	40	520	60	40
Fat	231	80	100	252	80	100	292	80	100
Lung	715	80	40	830	80	40	1080	80	40
Blood	1100	100	95	1200	100	95	1550	100	95
Disk	642	75	80	934	75	80	1210	75	80
Liver	425	45	70	500	45	70	641	45	70
Pancreas	550	70	65	600	70	65	780	70	65
Kidney	575	60	60	650	60	60	840	60	60
Spleen	662	60	70	775	60	70	1010	60	70
Vitreous humor	1600	100	95	1800	100	95	2340	100	95
Cardiac muscle	725	55	70	870	55	70	1114	55	70
Breast	425	50	85	500	50	85	650	50	85
Phantom fluid	330	300	50	360	320	50	450	380	50
Gado T1=150	150	70	95	150	70	95	150	70	95
Gado T1=100	100	55	95	100	55	95	100	55	95
Gado T1=50	50	30	95	50	30	95	50	30	95

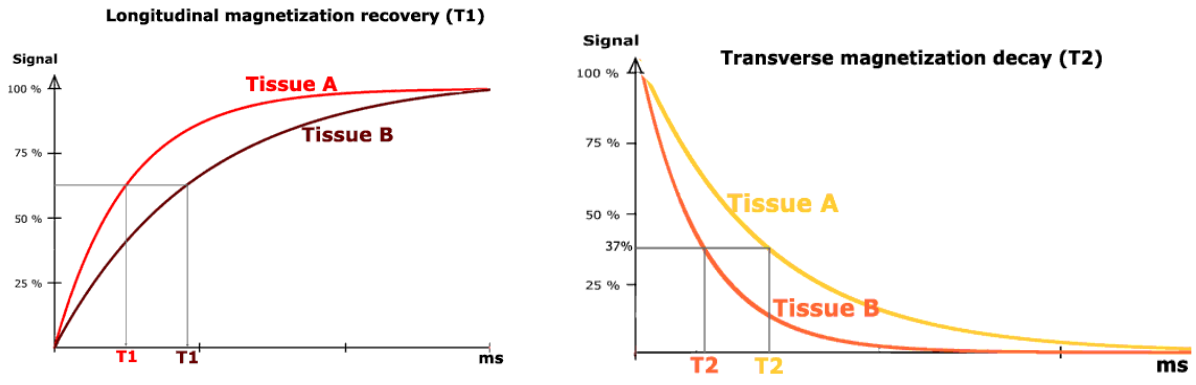


Figure 8: Decayment curves for different tissues. The difference in signal intensity between the two tissues at the specific time the signal is sampled is the bases from the image's Contrast and Weight. Extracted from [13]

The density of protons (PD) (also known as Rho) within a tissue is another factor that needs to be taken into account for the image's weight. Different tissues have different chemical compositions and as a result the number of water-bound hydrogen atoms and fat-bound hydrogen atoms present per unit of volume vary between tissues. A new image Weight, defined as Proton Density (PD) can then be obtained. In PD images neither T1 nor T2 contribute significantly to the overall signal/contrast-to-noise ratio (SNR/CNR). It is in fact, because neither T1 nor T2 have relevance on the sampled signal that the contrast based solely on the tissues' amount of hydrogen is discernible. Tissues with low concentration of water and fat-bound hydrogen will appear very dark (e.g. Ligaments) while tissues with high concentration of hydrogen will appear bright or very bright (e.g. Cartilage).

The high signal of the cartilage plus the very dark signal from ligaments that PD provides makes it, very often, the best Weight to assess joints (alongside T1). One of the drawbacks of PD Weight is that fat is also very bright, potentially causing glare effects in nearly tissues. To overcome this problem, Fat saturation (FAT SAT) (see subchapters: 10.2 and 10.3) is normally used in PD Weight. The result is suppression of the signal originating from fat tissues, leading to images with same clinical information as PD Weight but enhanced contrast.

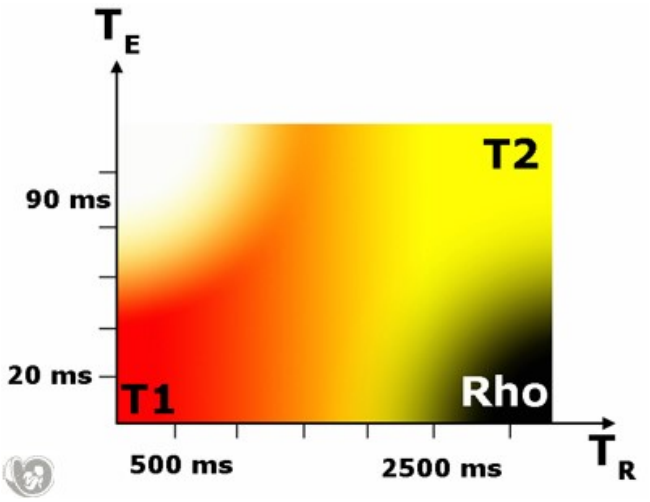


Figure 9: Obtainable Weights, at 1.5T, for SE sequences considering the selected TR and TE. Extracted from [13]

As exposed before (see subchapter 6.3), echoes can be created by both refocusing pulses and dephasing/rephasing gradients. The resulting decayment curves vary according to the method used for echo creation. As expected, this influences changes to the Weight of an image.

This is particularly true for T2 Weight when Gradient Echo (GE) sequences are used. The lack of proton refocus when echoes are created using gradients, instead of RF pulses, speeds ups the relaxation of the MT, resulting in a faster decayment that resembles the decayment curve of FID. The resulting effect is a significant change in image contrast that is considerably different for normal T2 contrast. To avoid confusion T2 Weight based in Gradient Echo is defined as T2 Star (T2*). It produces very high contrast between fluids and other tissues, even when compared with standard T2. It has very high sensibility to flow and susceptibility related artefacts/signal (e.g. Haemoglobin/hemosiderin deposits appear as metal artefact) and significant signal variance within a specific tissue

T1 Weighted images created using Gradient Echo methods do not differ as much from standard SE T1 images, yet loss of contrast between different tissues occurs and signal intensity heterogeneity within a specific tissue increases.

Defining the weight of the images (Figure 10) is done indirectly, by manipulation of the following basic MR parameters: TR, TE and Flip Angle (FAng) (see subchapter: 9.1.1)

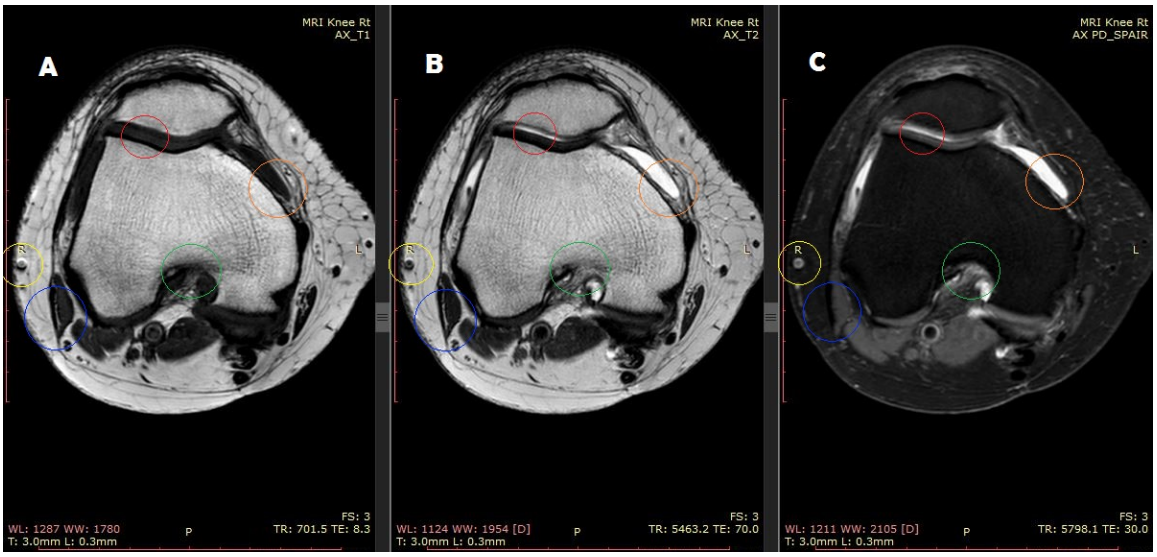


Figure 10: Axial MRI images of the knee in different image Weights: A - T1 Weighted; B – T2 weighted; C – PD Weighted with fat saturation. Different images' Weights allow the observer to characterize different tissue and structured by: first by matching the observed signal intensity of the structure/tissue, on the identified Weight of the image, against the expected signal intensity of the same structure/tissue in the observer's mental visualization/construct of the ideal image's Weight; and second by comparing the signal intensity of the structure/tissue against its signal intensity on other available image's Weight. The green circle demonstrates how the boundaries between different tissues can either be easily seen (C) or not seen at all (B) depending on the observed image's Weight. The Yellow circle demonstrates how Chemical Shift artefact is present in T1 (A) & T2 (B), but gets corrected in images with fat saturation (C). The Red circle demonstrates how T1 Weight (A) is unable to differentiate properly between cartilage and water compared to B & C. In addition it also shows that cartilage has much higher signal intensity in PD (C) Weight than T2 (B). The Fat saturation component of image C just further enhances the contrast between cartilage and other tissues. This is why PD_FS (C) is a very important image Weight for joint assessment. The Blue circle demonstrated how fat saturation can also be detrimental by reducing the contrast between fat tissue and other tissues that have low signal intensity (e.g. Ligaments) (C), compared to non-fat saturated images (A & B). Lastly the Orange circle demonstrates how image's Weight with high contrast can sometimes lead to a loss of image detail (A &C) in areas of very small/narrow structures, making Weights with softer/lower contrast (T2) better to assessment these areas.

8 MRI Sequences

There are several different sequences that can be used in clinical MRI. All of them differ from one another by either the way they excite the atoms, the way they encode the signal in space and/or the way they sample the echoes. All sequences have they usefulness, possessing both advantages as well as limitations/drawbacks comparatively with the other sequences. The key point is to know when to use them to achieve the desired image contrast and quality, considering the physical limitations presented (e.g. there is no point in using a SE STIR to acquire images of an abdomen since the images will simply have too much motion artefact to be useful).

Sequences can be organized according to the way they work, forming a "relationship" network (Figure 11) that allows for a better understanding of how each sequence relate to the rest and what Weights can be obtained with each sequence. There are two distinct "families" of sequences: Spin Echo, which are sequences that employ 180º refocusing pulse(s) to create an echo; and Gradient Echo, which are sequences that use only dephasing/rephasing gradients to generate echoes

Additionally a third family of MRI sequences can be assumed from sequences that possess both characteristics from SE and GE. These sequences are denominated as hybrid and employ both refocusing pulses and a dephasing/rephasing gradient within the same TR in order to acquire information extremely fast while keeping a low intrinsic Specific Absorption Rate (SAR).

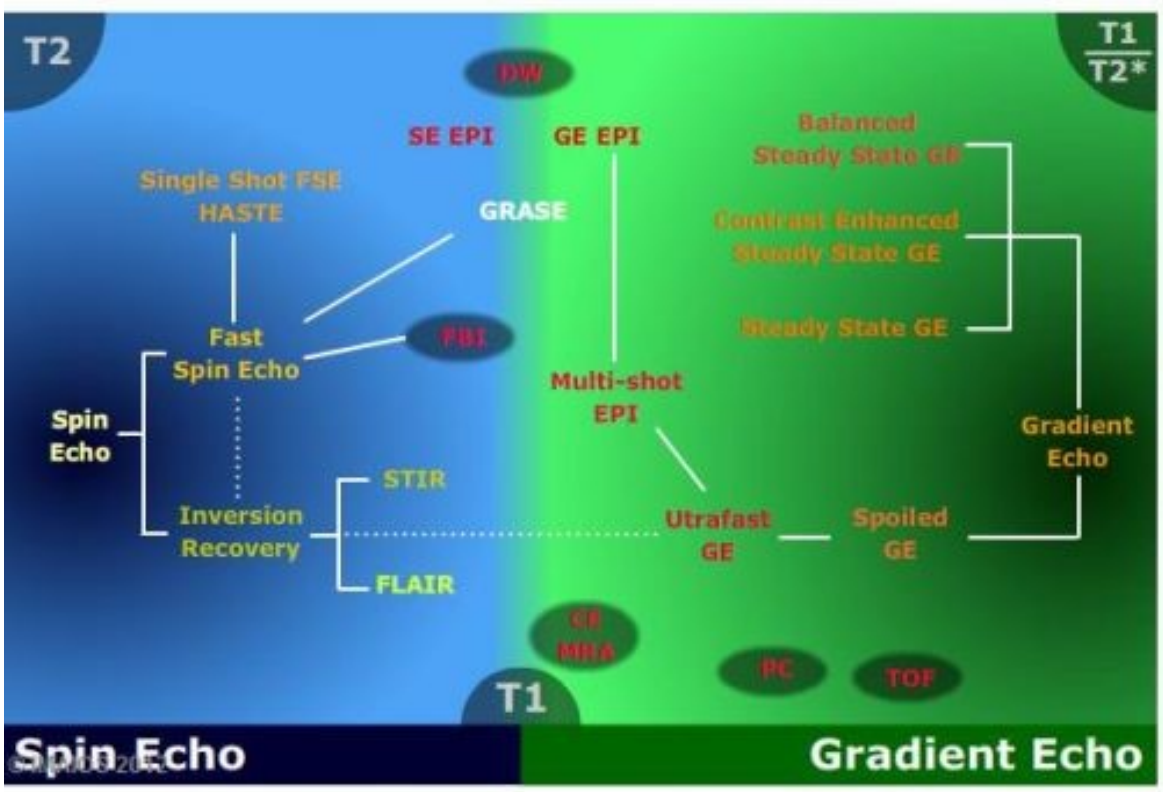


Figure 11: Diagram of MRI family sequences and their achievable weights. Extracted from [13]

8.1 Types of sequences used in orthopaedic MRI.

Since Orthopaedic MRI focuses mainly in assessing musculoskeletal pathology (Figure 10) of the appendicular skeleton the necessity of compromising image quality in favour of ultrafast scanning doesn't really exist, and even though motion related artefacts are far from uncommon in orthopaedic MRI, their presence and severity are nothing in comparison with thoracic-abdominal MRI and cardiac MRI. Simply put, because in Orthopaedic MRI, normally the area scanned is far away from anatomical structures that move significantly and autonomously, the motion artefact that such movement causes tend to not appear in musculoskeletal images.

The great benefit of not having to care (too much) about motion artefacts is that it allows for the use of sequences with greater acquisition time than the ones used on abdominal or cardiac scanning. By being able to afford greater acquisition time per sequence we can then focus more on increasing the resolution of the images and tissue contrast.

Ultrafast sequences tend to be inherently limited on the resolution, SNR and/or contrast of the images acquired and, generally, present themselves as not very flexible for parameter manipulation. So as a consequence of all these limitations, MSK MRI examinations tends to rely on sequences that take several minutes to acquire but that are able to obtain images with both high resolution and high contrast. What this means in practice is that in Orthopaedic MRI, the SE family dominates the spectrum of sequences used.

SE sequences present themselves as pretty much ideal for musculoskeletal studies, because they allow for acquisition of high resolution, high contrast images in virtually all Weights possible, with or without selective tissue saturation, while possessing good resistance to small amounts of motion and

susceptibility artefacts. Even 3D scanning is currently possible with some variants of SE sequences, making it more unlikely that SE will ever be dethroned in MSK exams.

8.2 Spin Echo (SE)

As stated above, Spin Echo refers to the creation of an echo through the use of refocusing pulses. This is the defining characteristic of SE sequences. Regardless of the presence of inversion pulses, multiple echoes or variable flip angles, if the mechanism used to create an echo relies solemnly on refocusing pulses, then it is a SE sequence. This is because the sequence possesses (to some extent) the same relaxation decayment characteristics.

Even though there are several variations of SE sequences (see subchapter 8.2 and its subchapters), they are all based on the standard Spin Echo sequence: where one echo is generated per excitation pulse/TR in the following fashion: [5, 7, 11, 13, 20]

After the application of the initial 90º excitation pulse the spins became fully excited and start dephasing freely for a certain amount of time after which a refocusing pulse (normally 180º) is applied across the y axis, causing the spins to flip to a mirror position from their original position (when the pulse was applied). The change in position across the y axis implies an inversion of the spins' phase angles and a change in the energetic estate that the spins were in. Low energy state spins (parallel alignment) switch to the higher energy state (anti-parallel alignment) and vice-versa [7, 10-12]

The pulses do not disturb B_0 and so the precession frequency of the spins doesn't change.

With the application of the initial excitation pulse the spins on different energy states start relaxing and dephasing away from each other, leading to a decrease of the MT. Applying the refocusing pulse doesn't change the direction of the spins relaxation, only their position is mirrored and phase coherence corrected, transforming what was originally dephasing into rephasing. It's from this approximation in phase between the spins in the two energy states that the echo is generated.[7, 10-12]

The phase correction produced by the refocusing pulse (and the lack of B_0 disturbance) makes the relaxation of spins not sensitive to field inhomogeneities and extends the length of the decay due to spin-spin constructive interaction. This is the reason behind the resistant of SE sequences to susceptibility artefacts, true T2 decayment (as opposed to T2*) and longer TRs/TEs.

SE possesses excellent SNR because the existent spin-spin interaction and full excitation of all spins increase the overall Magnetization Vectors (MT_0 and ML_0). [11, 13]

Comparatively with GE, SE provides images with better contrast between tissues, particularly on T1 images, and better contrast homogeneity within the same tissue. [11, 13] This is why SE is the preferred family of sequences in MSK imaging.

The most significant disadvantages are the long TA and the high amount of energy deposited unto the scanned object, leading to increased SAR (particularly in IR SE).[11, 13]

The time at which the refocusing pulse is applied is half of the TE value. TE is defined as the time lapse between the application of the excitation pulse and the moment when sampling is performed (ideally when the echo reaches its maximal amplitude). [11, 13]

8.2.1 Multi-echo Spin Echo

In multi-echo SE sequences, two or more echoes are created, within the same TR, to create two or more groups of images with different Weights/Contrasts. By applying two (or more) 180º refocusing pulses after the initial excitation pulse, at specific times, two (or more) echoes are created at the defined TEs. Each echo is sampled and the lines read are used to fill different K-Spaces. The PE gradient is applied right after the refocusing pulse (because the refocusing pulses do not affect the protons phases) resulting on the acquisition of the same line in the different K-Spaces. In this way, two (or more) images with the same TR but different TE are obtained at the same time (Figure 12).[5, 11, 13, 19, 25]

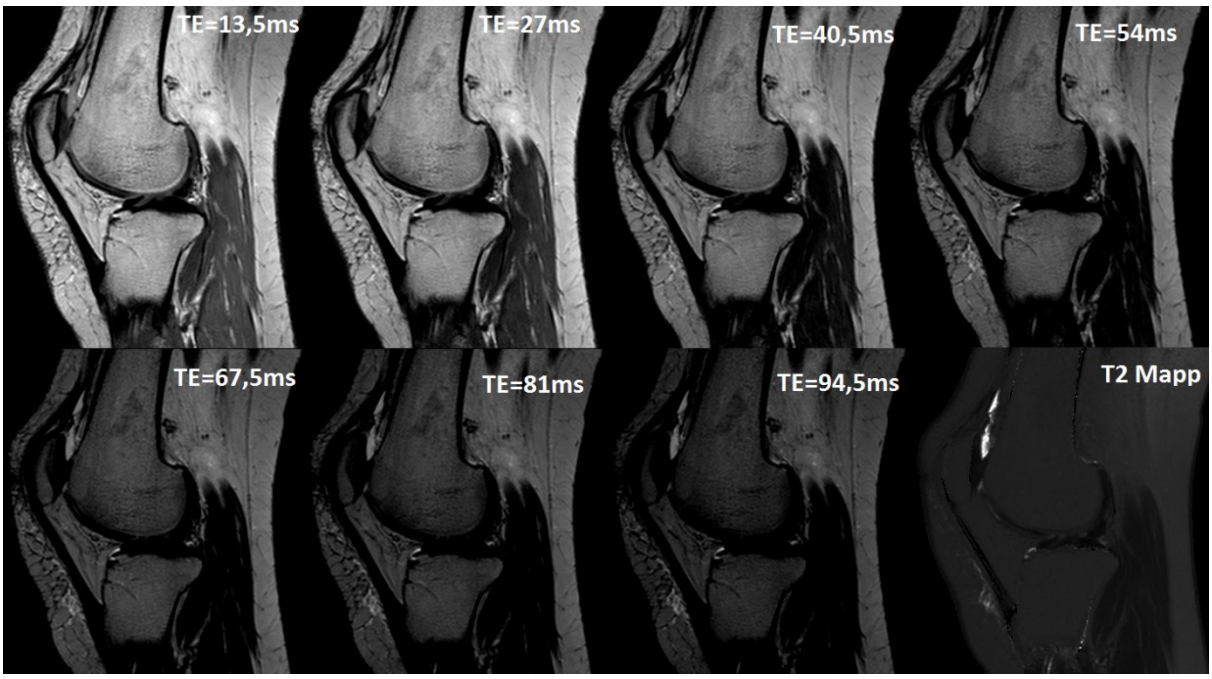


Figure 12: T2 parametric mapping sequence of a knee in sagittal plane. It's a multi-echo SE where TR=3425ms and 7 different TEs are sampled separately. It's possible to observe that the contrast changes gradually with increased TE, but that the Weighting also changes from a PD (shorter TEs) to a T2 (longer TEs). The decrease in the overall signal as TE increases exposes the MT decayment. Last image is a map, composited from the signal variance between the other 7 images for each tissue.

Since the only difference between K-Spaces is the TE, there is only a difference in contrast between the different sets of images (e.g. low T2 weight and high T2 weight). If the difference between the two TEs is very significant a difference in Weight can also occur (e.g. PD and T2)

The sequence's TR is fixed and only influenced by the longest TE defined; meaning that the sequence would take the same TA if it was a standard SE that had the same TR and the same TE (the longest one) as the multi-echo sequence. By this reason the images with the shorter TEs are often referred as "free images", because they were acquired without the cost of increased acquisition time.[5, 11, 13]

To accommodate for multiple echoes the TR tends to be long. T1 Weight is therefore not normally achievable with SE multi-echo. There is some significant freedom for the choice of TE values, so it becomes feasible to obtain PD images (long TR and short TE) at the same time as T2 images (long TR and long TE) (Figure 12).[5, 13]

As suggested earlier, images with the same Weight, but slightly different TEs are also possible, allowing for what is referred as "mapping" sequences where it is possible to assess how the signal of a specific structure (e.g. Cartilage) varies in time. Parametric mapping sequences enable the calculation of relaxation curves of the tissues (Figure 12).

The contrast obtained is identical to SE, with good sharpness, resistance to motion and susceptibility artefacts.[13]

Multi-echo sequences are not exclusive to the SE family of sequences. It's not only achievable with GE, but it's actually more commonly used with such types of sequences in MSK (see subchapter 8.3.1).

8.2.2 Inversion Recovery (IR)

From an acquisition point-of-view, Inversion Recovery (IR) sequences aren't very different from regular SE or Turbo Spin Echo (TSE), yet they provide a very distinct contrast.

They make use of a 180º inversion pulse, applied before the excitation pulse, to flip the position of the protons to a mirrored position from their original one (see subchapter 10.1). By properly spacing the time between the inversion pulse and the excitation pulse, the IR sequence allows for the saturation (see Chapter 10) of signal originating from a specific tissue (Fat, Water, Cerebrum-Spinal Fluid (CSF), Grey Matter, White Matter, etc.). Additionally, the contrast of the whole image is inverted, in comparison to a normal SE contrast, because after the application of the excitation pulse the tissues with longer T1 are closer to their original relaxed position than tissues with shorter T1 (M_L of long T1 tissues is closer to their M_{L0} than the short tissues M_L is).[13, 19, 25]

IR sequences allow for robust partial or full saturation of specific tissues, with average resistance to susceptibility artefacts, good resistant to B_0 inhomogeneity but very susceptible to flow and motion.[5, 13, 26] These drawbacks are a result of the long/very long TR that IR requires, flaws in the inversion/excitation process and in the case of flow artefacts the inflow of unsaturated fresh blood into the area covered by the FOV.

IR sequences can make use of echo trains to reduce TA, just like TSE (see subchapter: 8.2.3.1), becoming denominated Turbo Inversion Recovery (TIR) sequences. IR is not restricted to SE sequences. It is also very useful in GE sequences to achieve certain weights/contrasts.[5, 11, 27]

Multiple tissue saturation is also possible (e.g. Fat tissue and fluids). In this cases IR becomes Dual Inversion Recovery (DIR).[27]

Unfortunately, IR sequences possess inherently low SNR and a long/very long TR, making TA long and limiting the resolution of the images.[5, 13, 26] This is something that is somewhat attenuated by the fact that IR sequences have much more value for their very high contrast and sensitivity to show active pathologic processes than showing actual anatomy. IR and TIR are also more difficult to manipulate, comparatively with SE and TSE sequences, because they are less stable and major changes to the overall sequence can occur in association with small changes in the sequence's parameters.

The sequence diagram for echo creation, spatial encoding and sampling, does not change to accommodate for saturation of different tissues (unless multiple tissues are saturated), only the time between the inversion pulse and the excitation pulse (and forcibly the TR) is added. This time is defined as Time of inversion (TI) and it's specific for each tissue because it depends on the tissue's T1 value (see subchapter 10.1)

8.2.3 RARE

The single greatest inconvenient of SE sequences is their lengthy acquisition time. To reduce this limiting factor, variations of te standard SE where developed that deliver similar contrast while providing a very significant reduction in acquisition time. This group of SE variations is called Rapid Acquisition with Relaxation Enhancement (RARE).

8.2.3.1 Fast/Turbo spin Echo (FSE/TSE)

FSE/TSE sequences rely on the sampling of multiple echoes per TR, to enable the filling of more than one line of K-Space per TR. Being able to fill more lines of the K-Space in one TR, means that the total number of TRs needed to completely fill the full K-Space is reduced, which directly leads to a shortening of the TA.

$$TE_{echo}(ms) = number_{echo} * ES(ms)$$

Equation 3: TE of an echo of a TSE sequence.

To do all this, multiple echoes have to be created within the same TR. Several refocusing pulses (normally 180º, but not necessarily) are applied after the initial excitation pulse. Phase encoding is applied just after the excitation pulse because the phase of the protons remains the same, even after the application of the refocusing pulses. The multiple refocusing pulses form an Echo Train (see subchapters 9.2.1 & 9.2.2) and are separated in time evenly, making the echoes evenly separated in time as well (generating a new and important MRI parameter called Echo spacing (ES) (see subchapter 9.2.3). The TE value of the each of the multiple echoes will be equal to the number of the echo itself multiplied by the value of ES. [11]

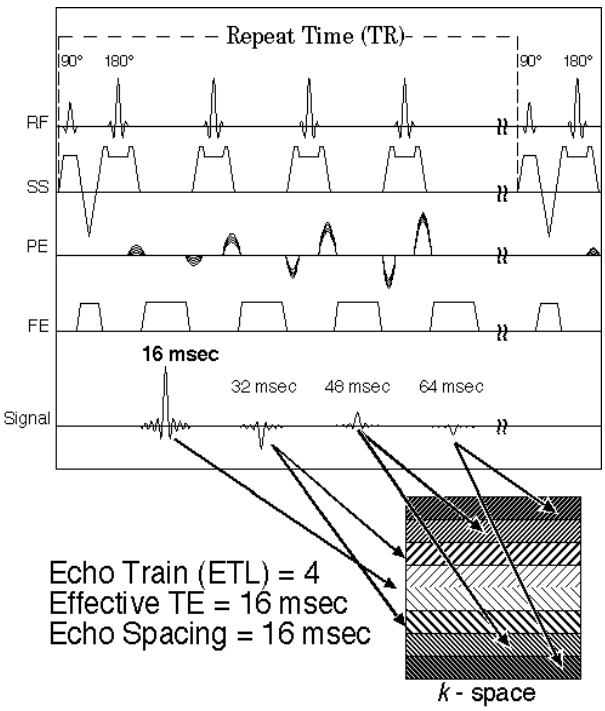


Figure 13: Fast Spin Echo sequence diagram and respective K-Space filling. Extracted from [9]

It's very similar to a multi-echo sequence but in TSE sequences all echoes contribute to a single K-Space (Figure 13). The result is that there is only one contrast for all images and this contrast isn't exactly the same as an SE sequence's contrast.[5] In TSE the contrast is a balanced mixture of all

the echoes sampled (known as effective TE), where the TEs of the echoes, that contribute to the lines that fill the centre of the K-Space, dominate over the ones closer to the edges of the K-Space (because it's at the centre of K-Space that information regarding the image's contrast is stored, while sharpness and motion information are kept at the edges (see subchapter 11.1)). By making sure that the centre lines of the K-Space are filled using the first echoes sampled (through selection of an adequate Profile Order (see subchapter: 9.2.7)) is it possible to avoid too much contribution from M_T to the image's Weight, allowing for acquisition of good T1 or PD images (depending on the TR).[5]

Some of the most relevant drawbacks from TSE are the potential for generating high level of SAR, potential contrast variation between slices and generating blurriness in the images when using long echo trains (see subchapters 9.2.1 & 9.2.2).

Additionally, the application of multiple refocusing pulses alongside short ES, leads to J-decoupling of fat bound hydrogen atoms. The result is an apparent lengthening of fat tissue´s T2 that appears in the TSE images as increased brightness for fat tissue, when compared to conventional SE images.[7, 11]

8.2.3.2 Single-shot Spin Echo

This sequence relies on very long echo trains (in the order of 100 or more echoes per TR) to obtain short TA while maintaining a spin-echo type contrast. The ultimate objective of this sequence is to allow acquisition one image per TR.[5, 7, 11, 13, 24-26]

The trade-off implies a significant blurriness of the images, high SAR and some reduction in image contrast. This makes single-shot sequences not suited for high resolution images, being instead more useful for abdominal scans, where the need to acquire allow of information within the time of a length breath hold in more important than resolution, or for localizers.[5, 7, 11, 13, 24, 28]

Single-shot sequences are often combined with Partial Fourier (see subchapter 11.1.1) K-Space filling (e.g. Siemens' Half Fourier Single Train Echo (HASTE)), to maintain quick acquisition while keeping the length of the echo train in check.[7, 13, 24, 25, 28]

Short TRs and short effective TE aren't possible due to length of echo train, limiting single-shot sequence to T2 Weighting only.

8.2.3.2.1 Half Fourier Single Train Echo (HASTE)

The full name for HASTE (trademark by Siemens) is Half-Fourier Acquisition Single-shot Turbo Spin Echo (other names apply for different manufacturers: USFE, SS-FSE, DIET. See page 139). It is used for sequential acquisition of low or medium resolution T2 Weighted images. TA is around one second per image. Echoes are generated using only 180º pulses. The Echo Train stars employing a high phase-encoding gradient. A Half-Linear or Reversed Half-Linear Profile Order is used to reduce blurring and improve T2 Weight of the images. Half-Fourier reconstruction is employed so only slightly more than half the raw data is acquired, reducing TA and SAR values.[25, 28]

Table 2: Standard SE Vs. Turbo SE Vs. Hybrid SE. Adapted from [5]

Characteristics	SE	TSE	GRASE
Contrast	High	High/Medium with fat signal appearing hyper intense	Medium/Low
Spatial resolution	High	High	Relatively high
SNR	High	Relatively high	Low
Sensitivity to motion artefacts	High	High	High
Sensitivity to susceptibility artefacts	Very low	Very low	Relatively low
Triggering	Yes	Yes	Yes
Reacquisition mechanism	Yes	Yes	Yes
Average scan time	Several minutes	2min – 6min	1min – 3min

Haste it's a particularly good sequence to produce localizers of spine, brain, and pelvis. Its SE basis makes it more robust than GE, when metal implants are present and it's quite adaptable for breath hold acquisitions.

8.2.3.3 Modified Spin Echo (MSE)

Modified Spin Echo is a variation of single echo SE sequences (available on Philips' MRI scanners) where a flip back pulse is automatically applied after the excitation pulse in order to speed up the realignment of the $M_{L,}$ from Hydrogen, atoms with B_0. This allows for TRs below 100 ms which enables shorter TAs and images with very high T1 Weight. [5, 26]

Image quality is improved compared to standard SE sequences with short TRs. [5, 26]

It is only compatible with standard SE and not any of its variants. [5]

8.2.3.4 3D TSE with variable Flip Angle (3D_TSE_vfl)

More commonly known for its commercial name: SPACE[1] for Siemens and Vista for Philips (see page 139), this type of sequence provides 3D acquisition with good SE contrast. [5, 28]

Unlike a normal TSE, 3D_TSE_vfl uses non-selective, short refocusing pulse trains (made of several pulses with different flip angles (FAng)), which allows for extremely high turbo factor (>100 echoes per train) and high sampling rates while keeping blurring effects partially in check. [28]

Even though the refocusing pulse trains are non-selective, the excitation pulses can be either slice-selective or slice-non-selective (depending on the operator's choice). [28]

This approach allows for 3D isotropic scans, with medium to high resolution in a reasonable TA (between 3min-7min) that can be freely reformatted post-acquisition while maintaining robust image quality.[5, 28]

Due to high turbo factor and subsequent long echo trains, short TRs are not possible. This means that T1 Weighted images aren't achievable. Yet PD, T2 and strong T2, Fluid Attenuated Inversion Recovery (FLAIR) and Inverted Recovery (IR) weighted contrast are obtainable with or without Fat/CSF suppression/saturation, making it a good option in most MRI examinations, particularly where T2 contrast and high resolution is necessary (e.g. Inner ear, Magnetic Resonance Cholangiopancreatography (MRCP).[28]

Normally three FAng modes are available (in Siemens' scanners), to allow for different Weights: [28]

- Constant: Uses constant FAng across echo train and a very long effective TE value, resulting in strong T2 or PD weighted contrast (depending on the TE chosen);
- PD variable: FAng is variable across echo train. Echo times are kept short to provide PD weighted contrast to the image;
- T2 variable: Variable FAng across ET with long TE, resulting in suppression of flow artefacts due to strong dephasing of the flowing spins. SAR is kept low. Allows for T2 and FLAIR weighted image contrast. For contrast comparable to T2 TSE it's necessary to use a longer effective TE.

The maximal number of slices possible to be acquired in one slab is limited to around 288. This makes the maximal coverage very dependent on the Volume Element (Voxel) resolution when using isotropic resolution.

[1] SPACE - Sampling Perfection with Application-optimized Contrast using different Flip Angle Evolutions

8.3 Gradient Echo (GE)

As exposed in earlier chapters, echoes can be created through the application of an additional pulse after the excitation pulse. Another way to create them is with the help of magnetic gradients used for encoding the signal in space. This type of gradient-based echo creation is the defining feature of GE sequences.

GE sequences require an initial excitation pulse (normally lower than 90º) and right after this a negative gradient (normally the phase gradient "Y") is applied (a negative gradient is nothing more than the act of the gradient creating a small magnetic field with opposite direction compared to B_0.) The negative gradient disturbs slightly B_0 field, inducing an accelerated dephasing of the M_T. The negative gradient applied is then reversed (becoming positive), forcing spins that were previously precessing at slow frequencies, to speed up and spins that were spin at high frequencies to slow down. This occurs because the tiny magnetic field created by the gradient that was previously "working against" B_0 is now "helping" it.[11, 13]

The inversion of the gradient also causes an inversion of the spins' direction of precession, which means that spins that were previously dephasing are now rephasing.[11]

Eventually, while rephasing, all spins will meet along the Y' axis and be in phase with each other, and then progress the dephasing as they move away from the Y' axis. The process of rephasing, meeting at one point (where they are all in phase) and then dephasing is what creates the echo (defined as the Gradient Echo). The echo is generated by the combination of each spins' M_T vector. When rephasing the vector value increases reaching its peak at the moment when all spins are in phase and then decreasing as the spins dephase from each other. [7, 11, 13]

Reversing the gradient only inverts the direction of the spin's precession, while maintaining the spins' initial phase (because the inversion of the gradient doesn't have refocusing properties), the echo produced is not the same as one produced with RF pulses (like on SE). An echo produced by gradients is much more sensitive to B_0 inhomogeneities, tissue susceptibility, flow, diffusion of protons and has no compensation from spin-spin relaxation. The result is an echo in which the signal's height is dependent on the FID's decay curve, which is dependent on T_2*, as the equation below exposes: [7, 11, 13]

$$S_{GRE} = S_0 \exp(-\frac{TE}{T_2^*})$$

Equation 4: Function of the signal's intensity of an echo created by a gradient. SGE = signal of GE; S0= initial signal height of the FID [11]

This renders T2 Weight impossible to be achieved, but instead T2* is obtainable.[25]

Additionally, the lack of phase refocusing when GE echoes are generated and the disturbance of B_0's homogeneity means that not all protons of the same tissue are in-phase with each other, producing significant signal/contrast heterogeneity within each tissue.

The FAng(s) used in GE to generate echoes are usually much lower than 90º. Smaller FAng (<45º) will produce T2* Weight. T1 can be obtained with a FAng higher than 45º. PD Weight can also

be attained, but this is very unusual in clinical MRI. Weight in GE sequences in also influenced by the chosen TR and TE, but the FAng as far more significance.[5, 11, 13, 25, 28]

By using FAng(s) lower than 90º, the protons are never fully saturated with GE sequences, giving them four identifying characteristics: [7, 11, 13]

1. Faster ML recovery, leading to shortening of TRs and TEs (in comparison with SE sequences);
2. Low contrast in T1 images, compared to SE;
3. Low intrinsic SNR;
4. Low intrinsic SAR.

In MSK MRI, GE sequences are much less commonly used than SE sequences, because overall the need for very short scans and low SAR is not as significant as the need to have high (and homogeneous) tissue contrast combined with high resolution and good magnetic susceptibility resistance. This doesn't mean that GE is useless in MSK MRI. Much to the contrary, GE has very specific roles in MSK, particularly for the assessment cartilage morphology, detecting pathology associated with hemosiderin deposits (e.g. Pigmented Villonudolar Synovitis) and dynamic examinations (e.g. Dynamic knee scan).

There are many variations of GE sequences; the majority of them don't have a specific purpose in MSK MRI. The most useful ones are exposed below.

8.3.1 Multi-Echo Gradient Echo

Just like Multi-Echo SE, GE Multi-Echo is a very similar form of its "mother" sequence (GE), but instead 2 or more echoes are created per TR.[25]

The multiple echoes sampled can then be used to create multiple images with different TEs, or be combined into a single contrast. In the first case creation of a parametric mapping it's possible; on the second case the sequence takes the denomination of Multi-Echo Data Image Combination (MEDIC – Siemens' commercial denomination (see page 139)). [25]

The combination of multiple TEs in the same image gives GE Multi-Echo inherent flow compensation, since the different TEs will reflect different degrees of flow artefact in different locations of the vessels. When all echoes are combined, the signal of the fluid in the vessels will be averaged out, becoming homogeneous throughout the entire image. MEDIC provides T2* contrast with high SNR (due to additional sampling per excitation, compared to regular GE). [25] It is particularly useful in MSK MRI for spinal and orthopaedic examinations.

8.3.2 Spoiled Gradient Echo

One of the results of very short TRs in GE (when TR<T2 of the tissues) is the presence of residual MT by the time that the next TR starts. This result in flipping of the residual MT by the excitation pulse of the second TR which leads to positive contribution from the residual MT to the signal sampled during the second TR.[13] This is known as Steady State and it has some interesting

applications in MRI, although it isn't usually employed in MSK imaging. When Steady State occurs, T2 Weight is enhanced in detriment of T1.[13]

To enable the acquisition of T1 images, the residual MT has to be destroyed before the start of the second TR. To do this a spoiling gradient (or RF pulse) is added to the generic GE.[13, 25] The intentional destruction of MT guarantees good T1 Weight and enables the use of shorter TRs and TEs.

A spoiling gradient works by randomly alternating, in each TR, the slice-selective gradient, while keeping the readout gradient constant.[13] Due to the lack of phase refocusing the alternation of the gradient disrupts any existing (continuous or intermittent) in-phase precession.

A spoiling RF pulse works by randomly changing the pulse's phase on each TR, preventing proper alignment between the multiple spin's vectors.[13]

Spoiled GE (e.g. Fast Low Angle Shot (FLASH)) advantages are low SAR, very short acquisition times and very good image sharpness, making them very ideal as localizers for anatomically complex areas, imaging using breath hold or other forms of acquisition triggers, and for 3D scanning. Contrast between tissues is very poor in comparison with standard SE.

8.3.3 Steady State GE

As mentioned on the previous subchapter, Steady State refers to state of pseudo stable precession, where there is a very small variation in FAng, throughout the entire sequence, and where MT is never fully lost between TRs. Under these conditions new types of echo, apart from the Gradient Echo, occur: Hahn echo and Stimulated echo.[7, 13, 25, 26]

Multiple types of echoes generate different contrasts, depending on which echo(es) are sampled during TR, and so multiple variations of Steady State GE exist.[7, 13, 25, 26]

Their use in MSK is very limited, particularly in non-contrast enhanced (CE) examinations.

8.3.3.1 Dual Echo Steady State (DESS)

DESS (Siemens's commercial name) is a specific variant of Steady State GE (see page 139). It consists in a 3D single-contrast non-spoiled GE sequence, with inherent fat saturation achieved through a water excitation method.[28]

Its main use is to assess joint cartilage, since DESS provides good contrast between cartilage and synovial fluid, plus the 3D allows for good post-acquisition reformats and segmentation.[28, 29]

DESS samples two echoes and acquires two distinct types of signal (Fast Imaging with Steady State (FISP) and Time-reversed FISP (PSIF) (see page 139)) within the same repetition time.[25, 29]

During image reconstruction, the strong T2 Weighted contrast of PSIF images is added to the FISP images. [25, 29]

It is especially good for 3D cartilage segmentation or for orthopaedic indications, where very good differentiation between synovial fluid and cartilaginous structures is essential. [25, 29]

Advantages of DESS: [25, 29]

- Improved SNR due to the acquisition of two raw data matrices;
- Strong T2 Weighted contrast;
- Acquisition time is comparable to a FISP sequence (see page 139);
- Excellent contrast between cartilage and other tissues;
- Allows for 3D acquisition under clinically acceptable acquisition times (at 3T).

8.3.4 Ultrafast Gradient Echo/Turbo Field Echo (TFE)

TFE is sometimes described as magnetization prepared GE in MR physics books or according to one of its commercial names (commercial name changes depending of manufacturer. See page 139). It is simply a basic Gradient Echo sequence, with very short TR (~6ms) and TE, preceded by a contrast-preparation pre-pulse. T1 Weight, in TFE, can be achieved by using an inversion pre-pulse (FAng=180º) or saturation pre-pulse (FAng=120º). By changing the preparatory period to a 90º-180º-90º set of pulses T2 Weight can be obtained. [26]

It's normally used in MSK for two dimensional (2D) or three dimensional (3D) dynamic studies (Figure 14) (e.g. Contrast wash-in and wash-out or patella maltracking) because the extremely quick acquisition enables capture of non-blurred images even during slow to medium movement of the scanned structures. Contrast is fairly poor.

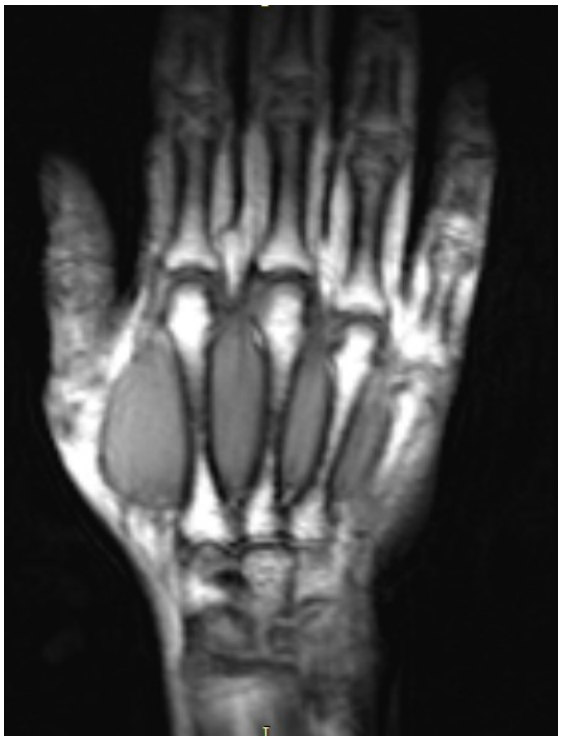


Figure 14: Coronal T1 TFE from a contrast-enhanced dynamic study of the wrist & hand.

1.1.1.1 Ultra-short Time of Echo GE

The TEs used in MRI are almost always greater than 1ms. Tissues that possess T1 and T2 times lower than 1ms will then always appear dark, because all the magnetization has already recovered to its stable state by the time sampling occurs.[27]

Currently there is a lot of interest in using sequences with TEs in the order of Nano and Picoseconds. Such TEs enable sampling of the signal decayment of tissues with very short T1 and T2 (e.g. layers of cartilaginous tissue, ligaments or cortical bone), providing insight into the physiology of such tissues and pathological processes related to them.[27]

Even though they aren't currently in use in the clinical setting it is expected that in some years Ultra-short TE sequences (Figure 15) will become a valuable asset in MSK, for the assessment of very specific body structures like the Enthesis, Intervertebral Discs, Meniscus and Cartilage layers.[27]

Producing sequences with Ultra-short TEs isn't as simple as just setting the TE value to its minimum and even after that is achieved one has to ensure that good contrast exists, between the tissues with Ultra-short decayment times and the other tissues in the vicinity (e.g. contrast between Bone Marrow and Enthesis). Hardware-wise Ultra-short TEs can be obtained at 3T or higher field strengths and require gradients with short slew times. Higher field strengths provide the high SNR required to be traded-off by TE reduction and short slew times allow for faster sampling which in turn allow for selection of shorter TEs.

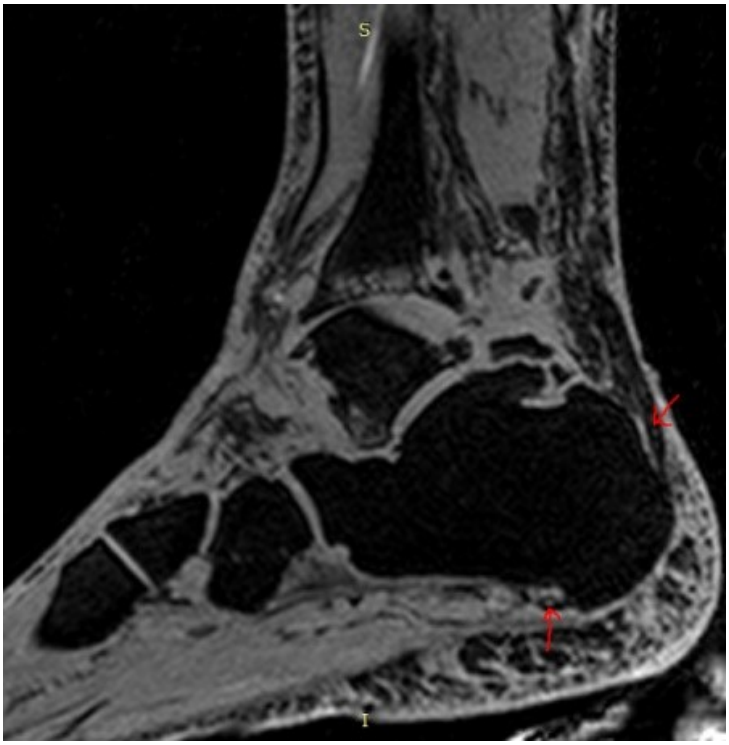


Figure 15: Ultra short TE Ultrafast GRE with fat saturation. TE=0.97ms. Hipertintense signal on the Enthesis is seen, contrasting with the hipointense signal of the ligament and the bone.

Possible Weights are still limited to T1, T1 fat saturated or Dual Inversion Recovery (fat and water saturation through inversion pulses)[27].

A TFE sequence with contrast preparation is essential to achieve Ultra-short TEs, alongside very high receiving BW, spoiling, very short FAng, RF pulse and Gradient modes set to maximum, very short ES and long/very long ET.

8.4 Hybrid Sequences

8.4.1 SE- EPI

It can be performed in the form of single-shot or multi-shot SE.[5] It uses Echo-Planar Imaging (EPI) (see subchapter 0) in conjunction with single-shot/multi-shot SE sequences to achieve extremely fast acquisition times of 2D images. Only T2 Weighted images can be produced.

EPI methods adjust well to single shot sequences because EPI requires very long echo trains, which are the intrinsic characteristic of single-shot SE. Limitations include very high sensitivity to motion, susceptibility and flow artefacts, low SE contrast, limited resolution, low intrinsic SNR.

Alternatively, and in order to reduce the limitations of single-shot SE-EPI, K-Space can be segmented in two or more regions, each one filled by information from one single-shot pulse. As a result sensitivity to flow and magnetic susceptibility are reduced and constricted to a limited area of the

image. Resolution is improved by reduction of the blurring effect, since the echo train used to fill each segment is shorter that the echo train used to filling the entire non-segmented K-Space/Raw Data.[5]

Lastly, the noise produced by this type of sequence is extremely loud as a result of EPI acquisition, which forces the gradients to work at maximum pace.

Table 3: Single-Shot SE-EPI Vs. Multi-Shot SE-EPI. Adapted from [5]

Characteristics	Single-shot SE-EPI	Multi-shot SE-EPI
Contrast	Intermediate	Intermediate
Spatial resolution	Low	Intermediate/High
SNR	Low	High
Sensitivity to motion artefacts	High	Intermediate
SAR	Low	Low
Noise produced	Very High	Very high
Triggering	No	Yes
Reacquisition mechanism	No	Yes
Average scan time	20s – 5min	1min – 5min (if cardiac triggering on)

8.4.2 GRASE

GRASE means "GRadient And Spin Echo" and reflects a sequence that combines characteristics of both SE and GE sequence. It is specifically a combination of Turbo Spin Echo (TSE) and Echo Planar Imaging (EPI). Its Contrast resembles more Standard SE T2 than TSE T2, but the scan time is shorter [26]

It has intrinsic low SAR but lower SNR, spatial resolution and lower resistance to magnetic susceptibility than TSE (Table 2).

It's used as an alternative to TSE T2 Weighted images in situations where SAR is a major concern and magnetic susceptibility isn't a concern for image quality(e.g. foetal MRI), but these kind of situations are quite rare in MSK MRI.

9 MRI parameters;

9.1 Basic parameters;

9.1.1 Image Weight, TR, TE and Flip Angle

As exposed earlier, in chapter 0, within MRI several types of image Weight are achievable, depending on how certain parameters are defined. These parameters define the way or time that the protons are excited, relaxed, saturated or sampled.[5, 7, 11, 19, 30, 31]

The main types of Weight are T1, PD and T2 or T2*. They express which magnetization vector dominates the overall signal's intensity of the tissues and the signal difference between various tissues. Every image contains some degree of contribution from the Proton Density, ML_0 and MT_0 but by properly setting the parameters we can define which one dominates other the others. If the parameters are properly chosen this dominance will be overwhelming, leading to a very good image weighting.

Traditionally, the relation between the different Weights and their primary affecting parameters (TR, TE and FAng) is described using subjective terms. This terms, like: short, long, low or high, are great in classroom physics but are fairly useless in clinical practice because they require a point of reference that doesn´t exactly exist. Additionally, these parameters are set, in the MRI systems, as numerical values not as qualitative/comparative terms, leaving the operator clueless about what exactly those numbers mean in relation to the desired Weight.

As an example, let's assume that the TR is 1000ms for a given sequence. Is it a long, medium or short value? The answer isn´t very straight forward, since 1000ms it's a short value for T2 or PD Weights but a long value for T1 or T2*.

There are several acceptable numerical values for every Weight, so a range of possible values for each parameter must be defined (Table 4). These ranges of values, for each parameter, must also consider other important factors that affect the relaxation curve of the protons.

Those important factors are as follow:

- Sequence (both family and type of variant);
- B_0 strength.

Ideally the ranges of values (Table 4), for each Weight, would be provided by each scanner´s manufacturer. Unfortunately that is not what normally happens.

Table 4: Range of value for TR, TE and FAng for the main/basic image Weights.

Weight /Type of Echo	TR values (at 1,5T) (ms)	TE values (at 1,5T) (ms)	TR values (at 3T) (ms)	TE values (at 3T) (ms)	Flip Angle (degrees)
T1 /SE	350-750	5-20	400-800	5-20	70-90
PD /SE	>1500	10-45	>2000	20-45	90
T2 /SE	>2500	70-130	>3000	70-130	90
T2* /GE	400-1000*	>15*	450-1100*	>9*	<40*
T1 /GE	<750*	<15*	<750*	<15*	> 50*
PD /GE	<750*	<15*	<750*	<15*	<40*

* Values for standard GE sequences. Values can change (normally are reduced) for other GE variants, due to the application of Spoiling Gradients, Contrast-preparation pre-pulses and the type of Echo(es) sampled. Always check the manufacturer's Sequence Manual for more precise information and/or cross-check the acquired images against the adequate literature.

9.1.1.1 Time of Repetition (TR)

Representing the length of time taken from pre-pulse preparation until the end of the echo train, the TR can be treated as the unit of time of any sequence. This is easy to accept when one understands that, time-wise, any sequence is just a specific TR repeated enough times (hence the name) in order to fill all desired lines of K-Space with enough true signal. Everything that occurs during a sequence occurs during the TR (Figure 16). It also becomes easily understandable that the TR, as a parameter, is at the core of almost everything in MRI and particularly Sequence Optimization.

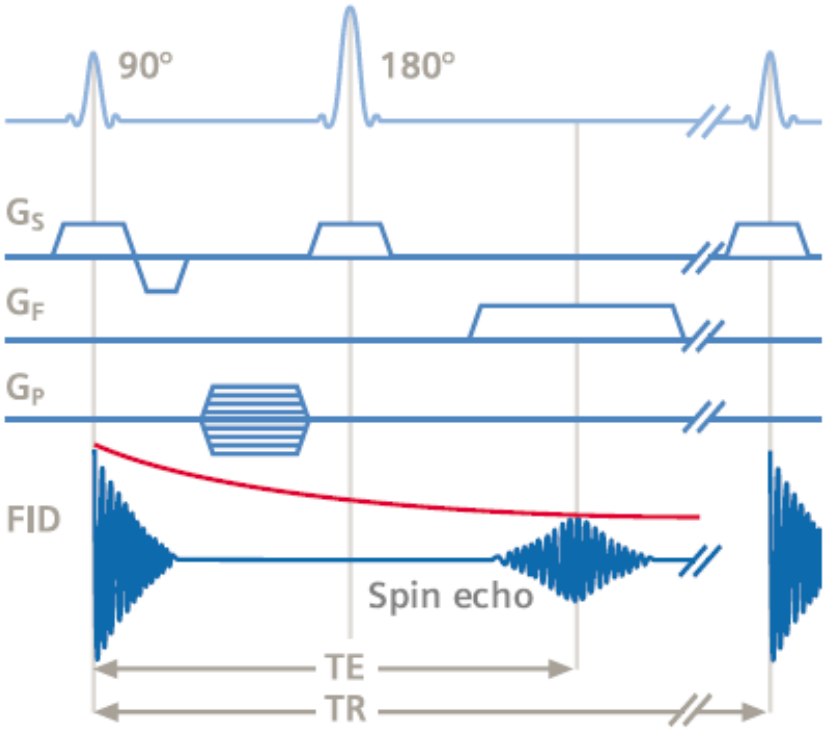


Figure 16: Generic diagram of a SE sequence.[3]

The large majority of parameters affect the acquisition time by changing the length of the minimum TR. On top of that, the chosen TR determines the number of slices acquired per TR, the contribution of T1 Weight to the overall Contrast and Weight of the images, and influences directly the SNR per Number-of-Excitations (NEX). This is why the TR should be first and last parameter to check before running any sequence.

Considering only the ML, and as exposed in Figure 17 & Figure 18, the overall intensity of the signal (and so SNR) increases, over time, in a logarithmic fashion, for both SE and GE sequences.

In an attempt to obtain the highest intrinsic SNR and the highest number of slices per TR, the longest possible TR (considering the Weight, type of sequence, and other contrast-affecting factors) should always be used. This is not a problem for PD or T2 images since a long TR is exactly what is required for good Weight. However, this can be problematic in T1 weighted images, where TR must be short. To counterbalance the loss of T1 contribution that occurs in a long TR, the FAng can be slightly reduced from 90º to 70º.[5] This strategy is particularly useful at 3T where the intrinsic higher SNR can be exploited to be traded-off for better contrast with good resolution in a slightly shorter scanning time.

Since the relation between TR and TA is linear (Figure 19) but the relation between TR and SNR is logarithmic, the ratio between SNR and TA varies depending on the chosen TR. This relation makes medium and long TRs preferable over both short TRs and very long TRs. This is particularly true for GE where SNR is more dependent on FAng and TE than on the TR itself.

Keeping a good balance between the length of the defined TR and the total number of TRs in a sequence is the secret for a time-optimized sequence. It´s crucial that the defined TR is as similar to the minimum TR as possible and that it is also within the range for the desired Weight.

Manipulating the number of Concatenations and Turbo Factor (TF) (see subchapters 9.2.4 & 9.2.1) with the aim of adjusting the TR to a desired range of values is a smart way to ensure that there is little to none wasted intervals of time, within the TR.

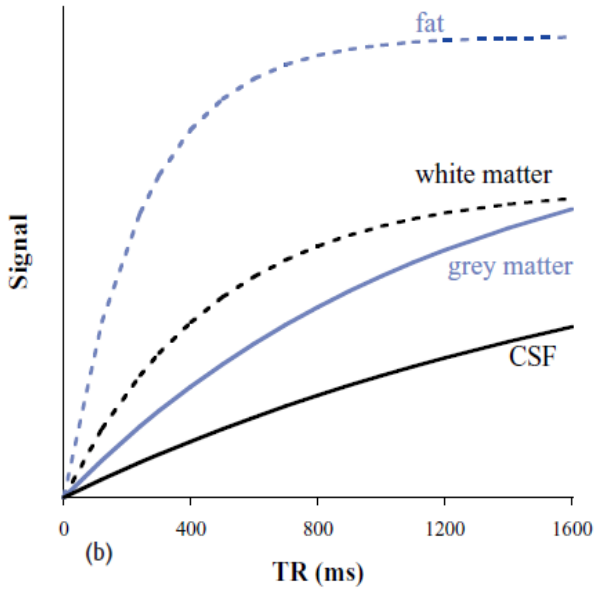


Figure 17: Relation between TR and overall signal in different tissues. Extracted from [1]

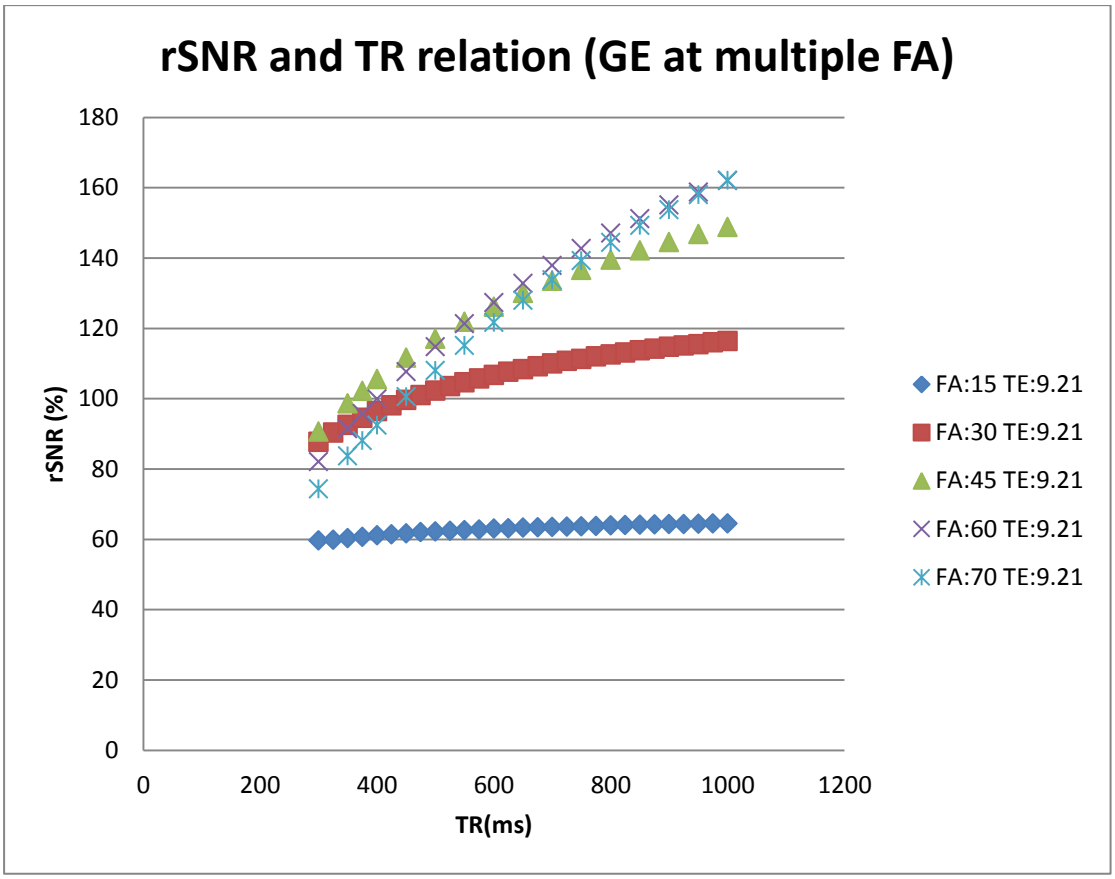


Figure 18: Relation between Relative SNR and TR in GE, considering multiple FAng(s). Data extracted from standard GE on a 3T Philips's Achieva X series. Reference sequence's parameters: TR=450ms; FAng=30º; TE=9.21ms.

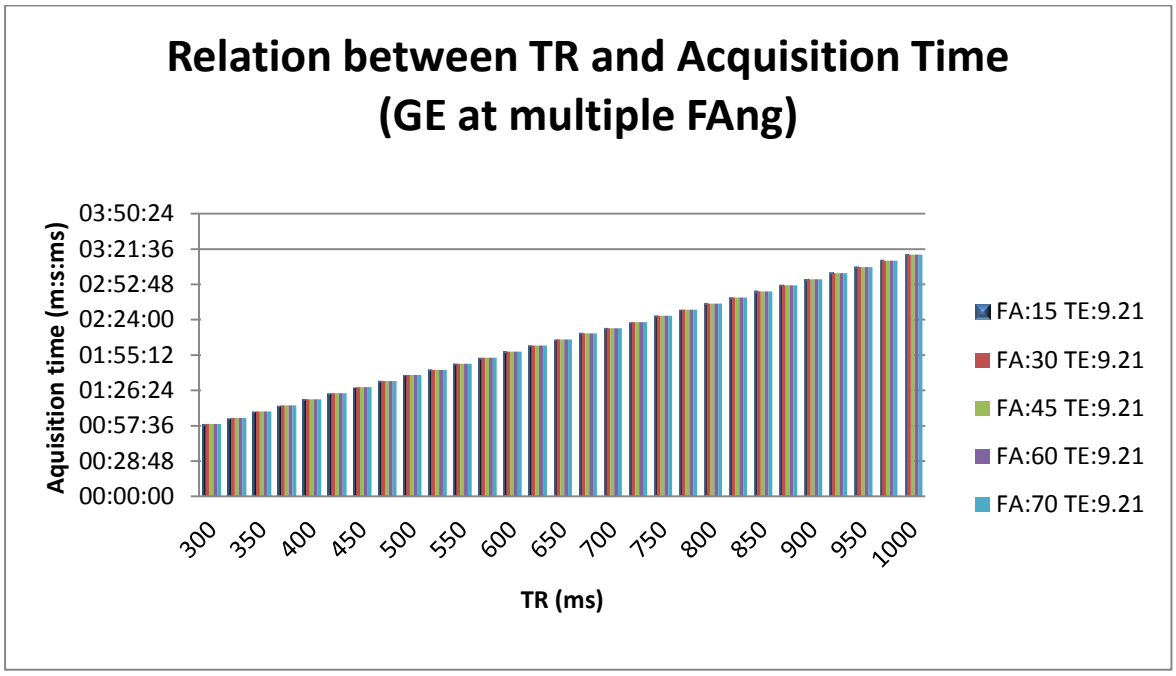


Figure 19: Linear relation between TR and TA on GE sequences at multiple FAng.

On a side note, if SAR levels are an issue, then the defined TR should be higher than the minimal TR. The time difference between the two TRs will be wasted time from an acquisition point-of-view, but that "useless" interval of time, will allow the patient to release heat to the surrounding environment preventing the build-up of energy within the patient and keeping SAR at acceptable values.

SAR levels are particularly relevant for pregnant patients, new-borns, and patients with metal implants (e.g. metal implant for scoliosis correction) or compromised thermoregulatory systems.

SAR is calculated just before each sequences starts. If post SAR calculation the SAR limits are exceeded, the sequence won´t run. Some systems (e.g. Siemens) will suggest changes to the sequence one wishes to run. Those suggestions will either be reducing the number of acquired slices, increasing the TR, reducing refocusing pulses´ FAng, or swapping to the next limit (higher) of SAR:

- Reducing the number of slices should never be selected, since it will change the area of coverage defined previously;
- Increase TR to the suggested by the system is acceptable if the suggested TR is within the range of acceptable TRs for the desired Weight. TA will increase accordingly;
- Reducing the FAng is acceptable only if the change isn´t very significant (up to 25º) (see subchapter 9.1.1.3) in order to not compromise the contrast between tissues. SNR will be slightly reduced;
- Swapping to the next SAR level won´t affect time, but the patient may experience some significant heat and start sweating.

Alternatively the TF, gradient mode or excitation pulse mode can be reduced (TR may need to be readjusted as the minimal TR will change).

9.1.1.2 Time of Echo (TE)

TE expresses the time interval between the moment when the RF excitation pulse is applied and the moment of the first sampled echo.[7, 11, 13, 18, 19]

This parameter defines mainly the contribution of MT (T2 Weight) to the overall Weight of the images. It also defines the moment when sampling occurs. Ideally the moment when sampling is performed should correlate with the moment when the echo reaches its peak (Figure 16), in order to guarantee the highest SNR possible.[5, 13]

TE is included within the TR, and so it´s always much shorter than TR. Just like the minimum TR, the selectable TEs (particularly the minimum TE) are dependent of the balance between other parameters: mainly the type of sequence, TF, ES, and receiving BW.

TE has as almost no effect on TA. It can only change TA by changing the length of the ET.

SNR on the other hand is significantly and directly affected by the defined TE (Figure 20) because of the hyperbolic fashion of how the MT decreases over time. For SNR's sake, TE is best to be kept as short as allowed for the desired Weight. This will also minimize susceptibility-related artefact, but improve the severity of flow effects. T2 contrast between tissues is usually greater at medium TEs.

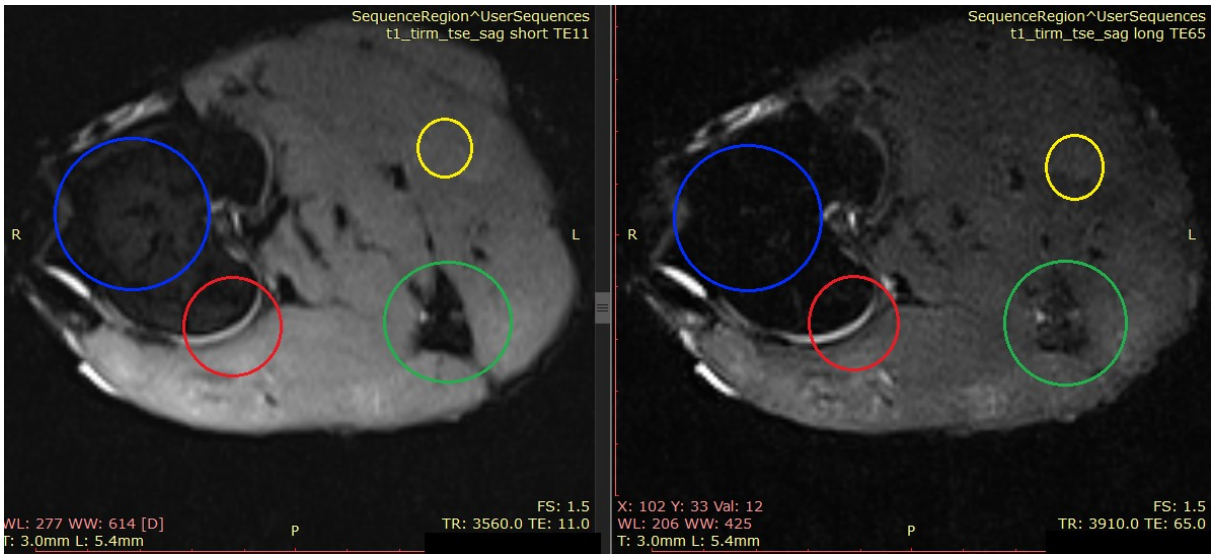


Figure 20: Axial STIR TSE (1.5T) of biological phantom at different TEs. As shown in the images above a change in TE from 11ms to 65ms can have drastic effects in many important technical aspects of an image: degree of Fat Saturation (Blue circle); Contrast (Red circle); Signal-to-Noise Ratio (Yellow circle); Contrast-to-Noise Ratio (Green circle).

As explained earlier, in sequences that use multiple TE(s) the time interval between adjacent TE(s) is defined as ES (see subchapter 9.2.3). In such instances the numerical values of the different TE(s) can be found using the following equation:

$$TE_n = TE_1 + n * ES$$

Equation 5: TEn = numerical value of the desired TE; TE1= numerical value of minimal TE; ES= Echo Spacing.

Depending on the MRI system used, the selected TE may refer to the TE sampled to fill the central line(s) of the K-Space or an average of the TE(s) sampled to fill the K-Space´s central lines (define as Effective TE).

Depending on the sequence used, the different TE(s) sampled can be processed in different ways, producing different results:

- If all TE(s) are combined in the same K-Space, then the sequence will produce images with only one Weight and Contrast defined by the TE of the central lines of K-Space. (e.g. Turbo Spin Echo);
- If very different TE(s) are processed in distinct K-Spaces, then the sequence will produce several sets of images with different Weights, exposing the same scanned area (e.g. Dual Echo Spin Echo);
- If slightly different TE(s) are processed in distinct K-Spaces, then the sequence will produce several sets of images with similar (not equal) Weight and Contrast, exposing the same scanned area (e.g. Mapping SE);
- If different TE(s) are processed in distinct K-Spaces, but then the resulting images are combined this will result in a single contrast that will be an average of all echoes sampled. (E.g. MEDIC).

In T2 images, a longer TE will produce higher contrast between fluid-like tissues (long T2 tissues) and other tissues, by enhancing their brightness and reducing the signal intensity of short T2 tissues.[5, 7, 11, 13, 24] In sequences that use Fat Saturation or Driven Equilibrium, the TE can be slightly shortened without detriment to the image´s contrast. The high contrast introduced by Fat Saturation or Driven Equilibrium compensates the loss of contrast that could occur with shorter TE(s). This is particularly pertinent in T2 Fat Sat weighted images, because a reduction in TE from a range of 70-120ms to a range of 60-80ms can significantly improve the poor SNR commonly seen in images with this type of Weight.

9.1.1.3 Flip Angle (FAng)

Reflects the angular variation experienced by the protons when a RF excitation pulse is applied. This variation is in relation to B_0´s/ML_0´s vector.[5, 13]

There are various types of flip angle, all of which force the Spin´s vector to tilt. Like RF pulses, FAngs are defined simply by the purpose of their application and the degree of tilting. The Refocusing flip angle, for example, reflects the flip angle observed when a refocusing RF pulse is applied.

In SE sequences the Refocusing FAng should be between 150º-180º in order to adhere to the Hahn condition and to not compromise the contrast between tissues.

In GE the Refocusing FAng is normally much lower than 90º. In this scenario, the Hahn condition does not exist leading to compromised contrast between different tissues, hence why GE sequences tend to have lower contrast compared to SE, particularly in T1 Weight.

For optimization purposes, manipulating the excitation FAng is less relevant issue in SE sequences because normally the excitation flip angle is (and should be) set to 90º, for full excitation of the protons. It can be slightly decreases to 70º on 3T field and higher, to improve T1 contrast when the defined TR is long.[5]

In GE sequences, this parameter gains significantly more importance (as TR loses it) in defining the contribution of T1 to the image. The Weight of an image is then defined by a combination of selected TE and Flip angle. As Table 4 exposes:

- T1 weight is obtained by combining a flip angle higher than 50º and short TE;
- PD weight is obtained by combining a flip angle smaller than 40º and short TE;
- T2* weight is obtained by combining a flip angle smaller than 40º and long TE.

Just like TR and TE, FAng is one of the fundamental parameters in MRI, affecting SNR, minimal TR (to a small scale), SAR and most important of all: image's Weight and Contrast.

As Figure 18 and Figure 21 show, the relation between FAng and Relative Signal-to-Noise Ratio (rSNR) is both complex and very relevant, exposing than a careful selection of FAng, considering the desired Weight, is fundamental to optimize SNR.

Associating the information in Figure 21 and Table 4 it can be concluded that for T2* and PD the flip angle should be as close to 40º as possible. For T1 it should be between 50º and 75º, increasing in direct relation with the TR used.

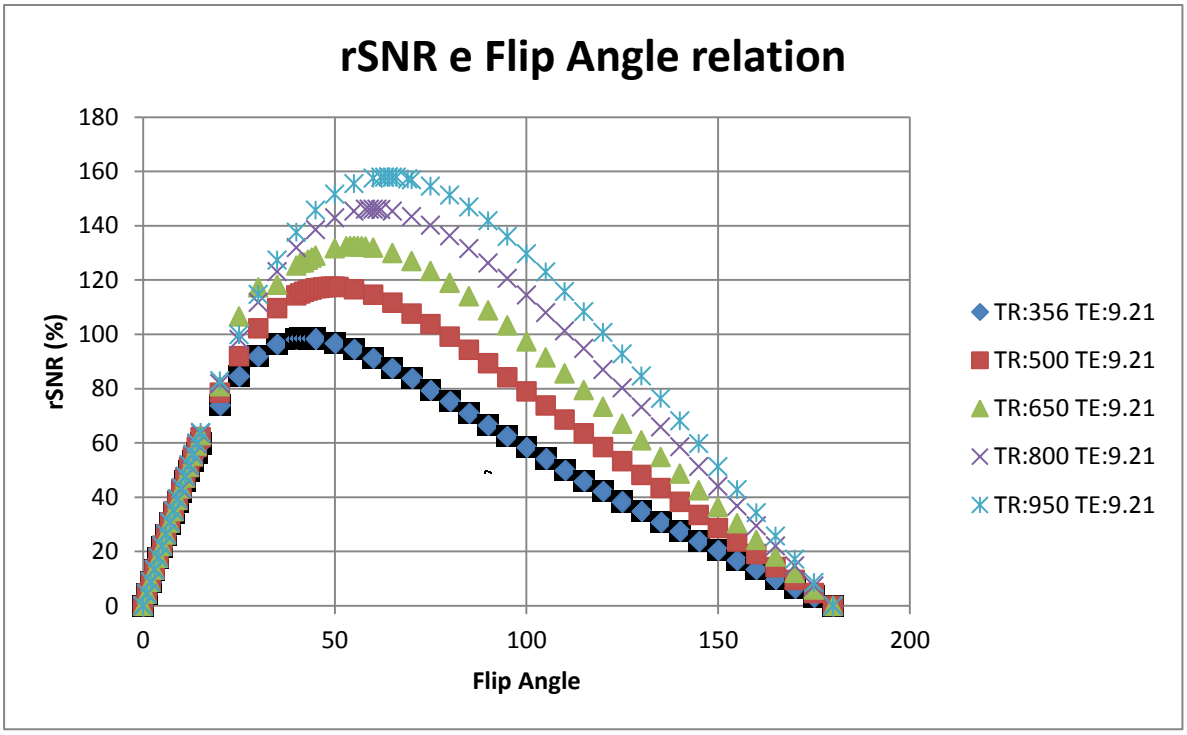


Figure 21: Relation between rSNR and FAng on GE sequences at several TRs.

SAR decreases with lower FAng because the FAng is a direct expression of how much energy is absorbed by the protons from the RF pulse/Gradient. This is why GE generates lower SAR (particularly at stronger B_0s) than SE to the point that it´s very unlikely that SAR limits can be achieved in clinical scanning with just GE sequences.

9.1.2 Phase encoding direction

When reading the MRI signal we need to encode it in space. To do so, three variable magnetic gradients are used:

1. The first gradient defines the slice/volume to acquire (normally defined as the Z axis);
2. A second gradient creates slight differences in the precession frequency of the in-plane protons, creating columns (normally defined as the X axis);
3. The third gradient creates a variation in phase between protons of the same column, creating the in-plane lines (defined as the Y axis).

Unlike CT or PET, these axes are completely arbitrary and can be interchanged according to the operator's desire.

The Phase Encoding Direction (Enc. Dir.), as the name states, refers to the direction in which the signal is read during the sampling period of the TR. This direction is always in the phase (Y) axis, and it is very important to consider during sequence planning because it is the direction where several kinds of artefacts show their presence. Flow, wrap/fold-over, ghost, breathing and streaking artefacts are

all very common in MRI and they always present themselves in the direction that the signal is encoded. To correct or minimize the appearance of the mentioned artefacts it's important to choose a Enc. Dir. That: it's parallel with the direction of major vessels (to reduce ghost artefacts); that is orthogonal to breathing movements (to constrain movement artefacts to small areas of the image that have no or few clinical interest); is parallel with the shortest in-plane axis of the object scanned (reduce or avoid wrap around artefacts).

9.1.3 Slice Thickness and Slice Gap

9.1.3.1 Slice Thickness

Slice Thickness (ST) is the effective physical width that one MRI image represents of the scanned object. It is related to the Z axis and is one of the most fundamental parameters to consider in the entire set of the parameters because it drastically and directly affects the SNR and indirectly relates to the TA of the entire sequence, by affecting the number of slices needed to cover the area of interest. Small changes in ST cause significant changes in the sequence's SNR, which can be traded-off (if needed) for significant changes in in-plane resolution, TA or Chemical Shift.

Significant though must be put into this parameter when creating and MRI protocol, because and if ill adjusted to the scanned body part or queried pathology, it can lead to significant partial volume artefact and reduce of image sharpness (if slice is too thick) or very long TAs and/or very poor SNR (if slice is too thin). Both situations can compromise severely the diagnostic quality of the images.

For areas of complex anatomy (e.g. Wrist), small structures/lesions (e.g. Fingers/Toes) or for detailed assessment of joints (e.g. Arthrograms), the Slice Thickness should be set to between 2mm and 3mm.

For examinations that use large Field-of-View s (e.g. Pelvis), large number of slices (e.g. Thighs), diffuse pathology (e.g. Myositis) or large lesions (e.g. Large sized Lymphomas/Lipomas), the ST should be between 3.5mm and 7mm. In both cases the thinner the slices the better, but image quality, TA and good in-plane resolution should be prioritised.

9.1.3.2 Slice Gap

It represents the effective physical distance between two sequential images. It is normally defined as a percentage in relation to the slice thickness. It's an area where, ideally, the protons are not excited, creating a physical barrier than prevents crosstalk between slices and the occurrence of partial volume artefact (Figure 22).

It should be kept as small as possible; to prevent potential pathological signs from being improperly covered.

Normally the Slice Gap is set as between 10 and 20% of the slice thickness. The bigger the slice thickness the smaller the Slice Gap should be (percentage wise), since a small percentage of a large thickness represents a significant nominal value of physical distance between slices.

In 2D sequences a gap smaller than 10% can lead to excitation cross-talk between adjacent slices, particularly if sequential slice acquisition (see subchapter 9.2.6) is used. There is no benefit in using a gap smaller than 10% to decrease the change of missing a small lesion. This being because even with 8mm slice thickness, 10% of gap represents 0.8mm distance between slices and sub-millimetric lesions are considered as not clinically relevant. In cases where the Slice Gap needs to be null or almost null, 3D sequences are preferable.

3D sequences usually have null Slice Gap, because the entire volume is excited simultaneously, but positive values for Slice Gap are also possible (either during image acquisition or reconstruction/reformations)

Unless there is poor slice selective excitation (e.g. Faulty gradients), increasing the Slice Gap over 20% provides no real benefit. It's preferable to maintain or reduce the Slice Gap and increase the Slice Thickness instead, since it will simultaneously improve SNR as well as increase the effective scanned volume.

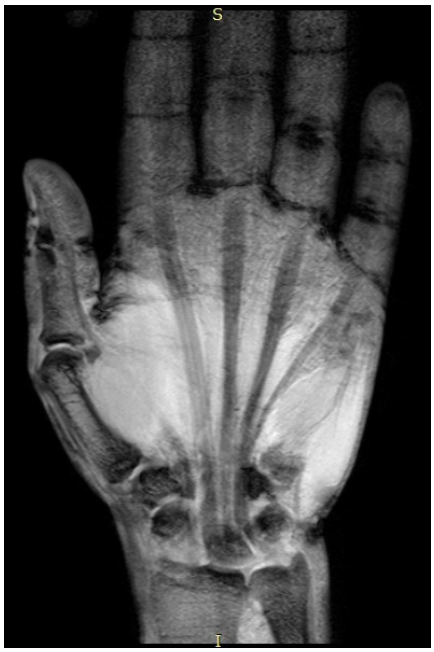


Figure 22: Partial volume artefact on Coronal T1 3D Spoiled Gradient Echo. Partial volume artefact is seen in this image as some tissues (ligaments) appearing semi-transparent. This is caused by the lack of slice gap in a 3D volume acquisition. In these sequences, during the excitation period, the Z gradient is positive along the entire volume, instead of just being limited to the slice that is going to be encoded.

9.1.4 Resolution: Field of View, Matrices and Pixels

The Field-of-View (FOV) consists on the physical area that will be covered during slice acquisition and will be exposed in the final image. During acquisition, the FOV is defined by the frequency gradient, through the time-amplitude integral [15]. The Matrix is a virtual grid laid upon the FOV dividing it into equality sized squares. The resulting squares are the image's Unit of resolution and since MRI images are digital in nature, these squares are called Pixels. The physical dimension (resolution) of an image's pixels is obtained by dividing the FOV's dimensions by the Matrix.

Traditionally the FOV is a square, reason why the resulting images are square-shaped. Some systems, however, (e.g. Siemens' and Philips' MRI scanners), allow the operator to re-shape the square into a rectangle with an area smaller than the initial square, which sometimes can be more useful to use. This change in shape is done by telling the system to not sample the outer lines of the matrix in the Phase dimension. This results in reduced SNR but also reduced acquisition time, because less K-Space lines are sampled. The amount of lines acquired is defined by the operator my manipulating the respective parameter ("FOV Phase" for Siemens' scanners). It's normally submissive to the defined FOV and expressed in percentage.

Some MRI systems (e.g. Siemens) use the FOV's dimensions as the point of reference for image resolution, while other systems (e.g. Philips) focus on the pixel's dimensions. Both methods work

in the same fashion, but behave differently when manipulated because of the interdependence between FOV, Matrix and Pixel.

To not create confusion, the following subchapters will assume that the resolution is dependent on the FOV and not the other way around.

9.1.4.1 FOV as parameter

Manipulating the FOV can have significant results in the overall balance between parameters. Due to the FOV's 2D nature, a small change in FOV causes significant increase in SNR and small decrease in resolution without affecting any other factors. This is why increasing the FOV is the most often and the simplest solution used by radiographers to ensure good SNR without changing the time of acquisition (TA).

Due to the degree of precision in resolution that MRI systems offer (between 0.1 and 0.05mm) it's possible to change the dimensions of the FOV a little, without decreasing the resolution perceived. To ensure that the maximal SNR for a given resolution is obtained, it is important to use the maximum FOV's dimension allowed by the chosen matrix.

The FOV must also be set in order to correctly adjust to the area of interest (e.g. when scanning a lump located in the left chest wall, only the left side of the chest should be scanned and not the entire thorax), and to prevent the occurrence of susceptibility, distortion and motion artefacts (e.g. susceptibility and distortion artefacts start to appear at the edges of the FOV, if the FOV is too large. This is particularly significant STIR, GE and any sequence that employs spectral fat saturation.).

9.1.4.2 Matrix as a parameter

Matrixes can be comprised of equal or unequal number of rows and columns. Usually the MRI system allows for definition of both dimensions of the matrix separately. The dimension referent to the Frequency Encoding Direction (N_{FE}) (also called Base Matrix) is normally the point of reference for the other matrix's dimension, which refers to the Phase Encoding Direction (N_{PE}). Just as the FOV, this last side can only be equal or smaller in number than the N_{FE}.

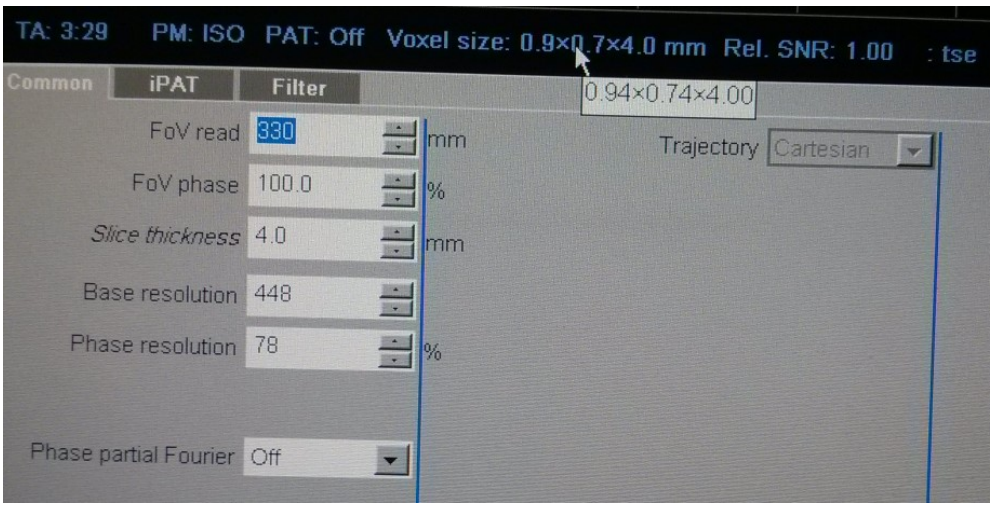


Figure 23: Siemens's Syngo MRI software: Resolution menu. Image showing degree of precision for voxel/pixel dimensions.

Just like when manipulating the FOV and for the same reason, there is some level of flexibility when manipulating the Matrix's dimension without affecting the resolution. (Figure 23)

The mathematical relation between FOV, Matrix and Resolution exposes why increasing either side of the matrix will cause a reduction of the pixel's size. SNR diminishes due to the fact that when the size of a pixel is reduced, the average number of protons contained within the voxel that such pixel represents also decreases.

Changing the Base Matrix with affect the dimension of the pixel in its two dimensions, providing around 20% more SNR each time the Base Matrix is reduced to the next size of possible matrices. Reducing only the N_{PE} of the will also improve SNR but to a significant less extent.

Time increases with higher matrixes, but in two different ways:

1. Although McRobbie, et al [11] affirms that changing the N_{FE} will have no effect on TA, in practice this isn´t exactly correct. A small increase in time is observed, through increase of the minimal TR, when the N_{FE} is improved. This is the result of a slight extension in time that the frequency gradient needs in order to achieve the desired amplitude, times the number of times the gradient is applied in each TR.
2. The increase in scanning time observed by improving the N_{PE} side of the matrix is much simpler to understand: For every extra phase line that needs to be sampled, time must be wasted in order to so. This increase in time is linear with the increase in the number of phase lines sampled and much more significant than the one observed for the N_{FE}.

Regarding optimizing the matrix parameters, it is best to keep the N_{FE} as high as allowed by the SNR (to ensure good resolution) and the N_{PE} as low as possible (between 65-100%), to boost SNR and keep the TA in check. The final resolution must ensure that the pixels are not too rectangular.

If the result is a low SNR image, then the N_{FE} should be decreased to the next lower value (and adjust TR accordingly).

9.1.5 Oversampling/Fold-over suppression

This parameter's name and behaviour varies between the different manufacturers, but its ultimate objective is to correct for the occurrence of unwanted fold-over artefact [5, 7, 11] (see subchapter 14.2.1.1.1).

In Siemens' scanners, this parameter is denominated as Oversampling and indeed consists in exactly that. The N_{PE} is extended behold the FOV so that signal for neighbour areas outside the FOV is also sampled. This extra-sampling improves SNR at the cost of time, because in reality more signal is sampled. By sampling the areas adjacent to the FOV is then possible to erase any fold-over artefacts from the oversampled signal, because the system knows that such signal isn't located within the FOV and so truncation without fold-over occurrence is much more likely. Like the Phase dimension of the FOV, Oversampling is set as a percentage of the FOV. The percentage of Oversampling is always equally divided by both sides of the FOV and can go from 0 to 100%. If the spatial resolution is quite

high, the Oversampling's maximal limit may be reduced due to gradient limitations. An Oversampling factor of 50% will be able to resolve any fold-over artefacts in the vast majority of cases. This parameter offers great flexibility in fine-tuning the relation between SNR and acquisition time, in contrast with NEX or IPAT, and doesn't compromise the balance between other parameters.

Philips' scanners on the other hand use a parameter called Fold-Over Suppression[2]. It cannot be fine-tuned, instead the only available options are: "On" or "Off". [5]

It operates in three possible ways depending on the number of Excitations (NEX) (see subchapter 9.1.6) defined by the operator:

1. If NEX=1, the scanner will produce two Sat Bands (see subchapter 9.1.7) adjacent to the FOV in the Enc. Dir. The two Sat Bands are not visible to the operator. This will increase the minimum TR and TA. Fold-over artefact is avoiding because the signal outside the FOV is supressed by the Sat Bands.[5] It may partially fail depending on the size of the FOV and the coil's area of coverage. This form of Fold-over is best for large FOVs;
2. If NEX=even numbers (e.g. 2; 4; 6; etc.), both the defined FOV and matrix will be doubled during acquisition (to maintain the spatial resolution) and the number of NEX halved.[5] None of these changes is exposed to the operator during planning. The reduction in NEX is performed to compensate the increase in TA and SNR produced by the FOV increase. After the acquisition is performed with a double-sized FOV, the data contained outside the initial non-double-sized FOV is assumed as having value equal to 0.[5] This form of fold-over correction based in truncation is more robust that Sat bands-based techniques. SNR and TA may still change because of limitation in the maximum FOV and matrix allowed by the scanner.
3. If NEX=3, both the both the defined FOV and matrix will be tripled during acquisition and the number of NEX reduced to 1.[5] It behaves much like when NEX is an even number. None of these changes is exposed to the operator during planning. No odd numbers above three are valid for NEX in Philips' scanners. [5]

9.1.5.1 Oversampling in volumetric (3D) scans

Oversampling in the Z axis (Slice) direction is also possible in volumetric scans (see subchapter 11.1.7). It works exactly like normal oversampling in the Phase direction, with exactly the same advantages and drawbacks, but instead of additional lines, "slices" are acquired. In volumetric scan some degree of oversampling (at least 33%) in both Phase and Slice direction should be performed to avoid fold-over artefacts, which can (and are likely to) occur in both directions.

[2] In the newest generation of Philips' MRI scanners this parameter has been substituted by Oversampling and it works exactly like Siemens' Oversampling parameters, with the added benefit that the amount of oversampling in each side of the FOV can be set independently for one another.

9.1.6 Averages/NEX

Averages or NEX, express the number of excitations needed to produce the desired SNR of the images. It literally represents the number of times that the entire sequence is acquired before reconstructing the images with all the acquired data. [5, 7, 11, 13, 18, 24]

NEX works in the same sort of way as applying layers of paint to a wall. If the paint has good quality (high intrinsic SNR) or if the wall is fairly smooth (low resolution), one layer is enough to evenly paint the wall. If the paint has poor quality (low intrinsic SNR) or if the wall is very rugged (high resolution), one layer of paint won't cover the wall evenly. In this last scenario more layers of paint need to be applied. Each layer takes time to be applied, so more time is needed to pain the wall (increasing NEX will increase TA linearly). Keeping with this idea of painting a wall but going the other way, if too many layers of paint are applied to the wall the amount of paint will be excessive. As a result the paint will start to run down and blur the sharpness of the final paintjob (image quality partly deteriorates).

A low value for NEX can result in grainy images, particularly in high resolution images (although many other factors also come into play in this matter). Low value of NEX will compromise both image quality and diagnostic quality if the degree of graininess is too significant. Excessive SNR, as a result of a high value of NEX, will produce images with a patchy/blotchy look that can also compromise the diagnostic quality of the images, although to a lesser extent.

Each increase in NEX is done at the cost of a significant increase in TA, but with the benefit of a substantial improvement to the overall SNR. This variation in both TA and SNR gets significantly smaller at higher NEXs. Each time the NEX is doubled, TA is also doubled while SNR is improved by $\sqrt{2}$ (around 41%).[5, 7, 11, 13, 18, 24]

To minimize the risk of motion artefact occurring as a result of long TAs and long examinations and improve workflow, NEX should be kept as low (normally around 1 to 2 NEX will be enough if other parameters are properly defined) as allowed by the SNR of the reconstructed images.

9.1.7 Saturation Bands

In some situations, it is advantageous to supress the signal originating from anatomical areas around the area of interest (either inside or outside the FOV). This may be because there is physiological motion arising from such areas that will produce artefacts in the area of interest or simply because there is high risk of fold-over artefact.

In such cases, Saturation Bands (also known as Sat Bands or REST (Siemens and Philips commercial names, respectively) can be of extreme help at the cost of a small increase in TA. Saturation Bands are rectangular regions of signal void (ideally) produced before the application of the excitation pulse, during the TR's pre-pulse phase. They are produced by selectively saturating the atoms within the defined region, making them non-responsive to the RF excitation pulse.

Just like slices, Saturation Bands can be freely manipulated by the operator, with no limit to the number of Saturation Bands, orientation and thickness.

With each Band the minimum TR is slightly increased, leading to TA increment in the order of 0s to 45s, depending of the sequence used.

The precision of the saturated region isn't perfect and gets worse the thicker the Saturation Band is and less effective with longer TRs.

9.2 Advanced MRI parameters

9.2.1 Turbo Factor

Turbo Factor (TF) defines the number of echoes produced per TR [5, 24] and its main advantage is reducing TA by diminishing the number of TRs needed to sample all K-Space lines needed. TF directly affects the minimum TR because the larger the number of echoes produced the longer TR and Echo Train (ET) will have to be in, order to accommodate for the extra echoes. By itself the TF does not introduce to much change to the sequence. It only does by influencing TR and ET, but the result of such influence needs to always be carefully pounder because the lengthening of TR and ET through TF increase is almost always quite substantial and sometimes it may present no benefit at all, even including the TA reduction.

The relation between TF and TA is not linear and can actually become detrimental for long TFs. It is therefore important to choose the adequate TF for each sequence. Such value is normally found by trial and error.

The adequate TF to use varies with the desired weight and the chosen profile order (see subchapter 9.2.7).

Experimentally the optimal ranges of TFs, at 1.5T, for MSK sequences are the following:

Table 5: Optimized TF values for different Weights. Data obtained with a 1,5T Siemens Symphony with TIM technology. Profile Order assumed: High-Low.

Weight	Optimal TFs
T1	4-7
PD	7-10
T2	12-18

If radial acquisition is used, then TF will also determine the number of lines per blade.[24]

Manipulating TF in conjunction with Concatenations is a great way to manipulate TR in order to obtain the desired weight in the best TA, but the effect on ET must also be considered.

9.2.2 Echo Train

It's defined as the time interval between the first and last echoes in a single TR.[5, 7, 13, 24] Its length directly influences the degree of image blurring and the fat's signal enhancement that arises from J-decoupling.

Large ETs produce significant blurring, reason why 3D sequences and sequences that use EPI acquisition produce blurred images, which degrades resolution in the encoding direction(s).

It's often hard to know how long the ET can be, before the blurring effect becomes significant. This happens mostly because some scanners do not provide the effective length of the ET, the fact that the limit varies with the type of sequence used and the lack of 3rd party standardization. Once again it boils down to a trial and error approach and to what degree of blurring the operator deems acceptable.

Table 6: Visual effects of multiple ET lengths on TSE's images. Based on images produced in 3T Philips Achieva.

ET Length (ms)	Degree of blurriness
<70	○ Very sharp images; ○ Small structures can be seen with good detail.
70-100	○ Sharp images; ○ Small structures can be seen with good to average detail.
100-135	○ Partially blurred images; ○ Borders between tissues become dull; ○ Small structures become hard to see.
>135	○ Partially or completely blurred images; ○ Blurred borders between tissues; ○ Contrast can become compromised: patchy appearance; ○ On the best cases small and medium size structures may be detected but solely based on contrast.

Image blurring isn't something than simply burst out of the nowhere as soon as the ET reaches a certain time length. It is something subtle that creeps up with ET lengthening. It initially obscures very fine detail (e.g. Trabecular bone), then eventually progresses into dulling of the borders between tissues and finally generates distortion of structures and contrast.

In TSE, which is the most common type of sequence used in MSK, ETs shorter than 70ms will produce extremely sharp images. Between 70 and 100ms, images will still be quite sharp and trabecular bone can still be discerned to some extent. When ET's length is within 100 and 135ms, the blurring effect may become perceivable (through the inability to see trabecular bone and dulling of tissue borders) but not significant enough to compromise the diagnostic quality of the images. Beyond 135ms of ET length, image blurring will likely become excessive.

9.2.3 Echo Spacing

It consists in the temporal interval between two adjacent echoes and is defined by the temporal interval between the applications of two adjacent refocusing pulses. It is determined by a relation, on a given sequence, between Turbo Factor, initial TE, Profile Order, RF pulse mode, Gradients mode and receiving Bandwidth.[5]

In some instances, ES can be defined, by the operator, "independently" of the selected TE (see subchapter 9.2.7.4).

After the initial TE (which is not the necessarily the effective TE of the image) is set, the ES defines the subsequent TEs of the echo train in the following manner:

$$TE_n = TE_0 + n * ES$$

Equation 6: Equation to calculate the value of a specific TE on a echo train. TEn – value of the desired TE; value of first TE of the echo train; n – number of the desired TE within in the Echo train; ES – Echo Spacing.

ES should be kept as short as possible to reduce the blurring effects of ET (by reducing the effective length of the ET), reduce the contrast difference between lines of the same image and reduce susceptibility artefacts.[5, 7, 11, 13]

The major concern with short ES is the signal enhancement that occurs in fat tissue. When the ES between two adjacent echoes is short a molecular phenomenon known as J-decoupling occurs. This phenomenon results in the lengthening of T2 decayment for fat tissue which is expressed on the images as signal enhancement over other tissues. [5, 7, 11, 13]

The most common detrimental effect of enhancing fat tissue's signal, on MR images, is an unbalance of both contrast and brightness, throughout the FOV, rendering the image visually harder to assess. It's more easily seen on Multi-Echo SE at long ETs, when scanning patients with very thick hypodermis and on Metal Artefact Reduction Sequence (MARS) images.

9.2.4 Concatenations/Packages

It defines how many times the TR should be divided to better accommodate several groups of slices, in multi-slice scans. In conjunction with TR, the Concatenations parameter (Siemens' nomenclature. Named Packages in the case of Philips) defines in how many single sequential measurements the slices will be distributed by. [5, 32]

It can only affect SNR by changing TR. [5, 32]

Increasing the number of concatenations will lead to increase TA if TR is maintained, because the difference between defined TR and minimal TR increases.

Concatenations is best used in partnership with TF in order to obtain the best TA out of the desired TR (considering the number of slices), or to attain low TRs, in multi-slice acquisition, when a large number slices are acquired in a single sequence.

When Multi Breath-Holding is enabled, the Concatenations parameter may also define (along with other sequence factors) how many breath-hold intervals the sequence will be divided (if this interval cannot be defined independently by the operator).

In IR sequences, higher number of Concatenations can lead to reduced sensitivity for inflow-artefacts by widening the slice profile of the inversion pre-pulse.[5, 32]

9.2.5 Driven Equilibrium (DRIVE/RESTORE)

Signal enhancement is not always detrimental. Deliberately enhancing the fluid's signal can be of diagnostic and technical interest.

This can be done by adding a refocusing pulse followed by another out of phase pulse at the end of the echo train. This combination of pulses at the end of the ET is called Driven Equilibrium and it significantly increases the length of the fluid's MT, which in turn improves fluid's signal.[5, 7, 11, 24, 33, 34]

The application of Driven Equilibrium results in the increase of ET's length, increase of the minimum TR, and also improved SNR. [5, 32-34] The SNR increase is particularly true for short TRs [32], where intrinsic SNR is more dependent on TR variation that TE variation.

Rudolf Radlbauer et all [33] reports that adding this pulse to the end of the echo train didn't affect the acquisition time, and Wilhelm Ruempler [34] is more measured, referring that "The RESTORE version of the T1 measurement did not cause any significant increase to the measurement time." Applying DRIVE/RESTORE does indeed affect the time of the sequence (because it increases the minimum TR) but in different amounts depending the value of the TF employed and the MRI system used. Generally speaking the acquisition time of the sequence will be increased by 5s-20s for T1, 15s-45s for PD and 15s- 45s for T2.

On T1 images, the application of this pulse to the overall sequence generates a variation in the T1 contrast, defined as T1 DRIVE or T1 RESTORE (commercial names for Philips and Siemens respectively). In this instance all tissues maintain the same T1 contrast, with the exception of fluids which changes their signal intensity of dark grey to light grey. [5, 32-34] This "whitening effect", as referred by Wilhelm Ruempler [34], on the image's contrast is similar to an increase in TE from 11 to 22 or 33ms, but with the added bonus of improved SNR, instead of the obvious decrease that this change in the TE value would cause.

The change in signal intensity allows for better discrimination between fluid and cartilaginous tissue, enabling better delineation of the cartilage and detection of fluid collections within the menisci and other cartilaginous structures. For this reason T1 DRIVE contrast is preferable over regular T1 in the evaluation of most joints.[5, 32-34] This is not the case when evaluating masses, as the contrast change may mislead radiologist. Additionally T1 images are less susceptible to artefacts compared to PD with fat saturation, which is the standard sequence for assessing joints.[33, 34] This is especially true for flow artefacts.

Since SNR is significantly increased in short TR sequences with DRIVE, other parameters may be reduced to keep the time of acquisition in check.

DRIVE can also be employed in PD or T2 Weight with or without fat saturation, making DRIVE much more useful from a technical point-of-view. When applying DRIVE to both PD and T2

images, TE can be reduced without detriment to the image's weighting/contrast. With DRIVE, good T2 contrast can be achieved with TE values as low as 60 - 65ms. [5, 24]

When this strategy is employed SNR will be improved, due to TE reduction. This is particularly effective on T2 Fat Sat images, where it can often be quite difficult to obtain good SNR within a satisfactory time of acquisition (without severely compromising the resolution).

9.2.6 Sequential vs. Interleaved acquisition of a group of slices

During excitation in 2D acquisition, the slice gradient is turned on to select not just the thickness of the slice but also which slice(s) are going to be acquired (excited) in a specific TR. This can be done in two ways: sequential and interleaved.

Sequential acquisition follows a numerical order (1,2,3,4…) for slice acquisition and can be done in ascending (1,2,3…) or in descending (…3,2,1) order.[35]

Interleaved acquisition uses an ascending order but acquires first odd number slices and then even number slices (or vice-versa).[5, 35]

The method of slice acquisition is important because of imperfections in the mathematical methods used to truncate the RF pulses used in MRI. The physical impossibility for gradients to have slew times equal to zero and the need to acquire several slices per TR (to reduce acquisition time), make it easily possible that information from one slice may be incorrectly encoded in the adjacent slice. This is defined as cross-talk between slices and depends on the quality of the slice profile selection (by both slice gradient and transmission pulse) and the distance (both physical and temporal) between acquired slices. The closer in space and time the slices are, the more chances exist for cross-talk.[5, 35]

Even though MRI scanners allow the operator to select between the two orders of slice acquisition, it is always advisable to select interleaved acquisition, as it presents generally no real disadvantages oven sequential slice acquisition and minimizes cross-talk. [5, 35]

Yet there are sequences that can only be done using sequential acquisition. In those cases the operator needs to adjust other parameters to reduce the occurrence of cross-talk. When acquiring slices at very specific times then sequential acquisition may be preferable.

9.2.7 Profile Order

Profile Order refers to the way in which each echo within an ET corresponds to a given line in the image's K-Space, by defining the order in which the lines of the K-Space are filled. This is actually quite important to consider because it defines the effective TE of the image (and so its Weight/Contrast) and it may also influence the ES for a given TE.[5]

This parameter cannot be manipulated by the operator in Siemens's scanners, but Philips' scanners on the other hand provide some very different choices.

The recommended Profile Orders for different Weights are shown in Table 7.[5]

Table 7: Recommended Profile Orders for different image weights, TEs and TSE Turbo factor. Extracted from [5]

Effective TE	Weight	Profile order	TSE Turbo factor
Long (90ms-150ms)	T2	Linear	High (10-25)
Very long (>300ms)	Strong T2	Linear	Very High (>30)
Short (10-50ms)	PD	Asymmetric	Intermediate (9-16)
Short (10ms-25ms)	T1 or PD	Low-High	Low (around 4)
Short (10ms-25ms)	T1 or PD	Linear with Halfscan (Half Fourier)	Higher than Low-High

9.2.7.1 Linear Profile Order

Echoes and K-Space lines are correlated in a linear ascending fashion (Figure 24), with the first echo corresponding to the bottom line of the K-Space, the middle echo to the central line and the last echo to the top line. It is suited for long effective TEs. [5]

The shot length is approximately 2*TE and ES equals: $ES = (2 \times TE) \div (TF + 1)$. [5]

If Partial/Half Fourier is combined with Linear Profile Order (Figure 25), it is possible to obtain a short effective TE but the sequence becomes very sensitive to blurring.[5] It this condition, K-Space is filled in the same order but ignoring the bottom lines of the K-Space, resulting in a change of the effective TE to a lower value than the one without Partial Fourier.

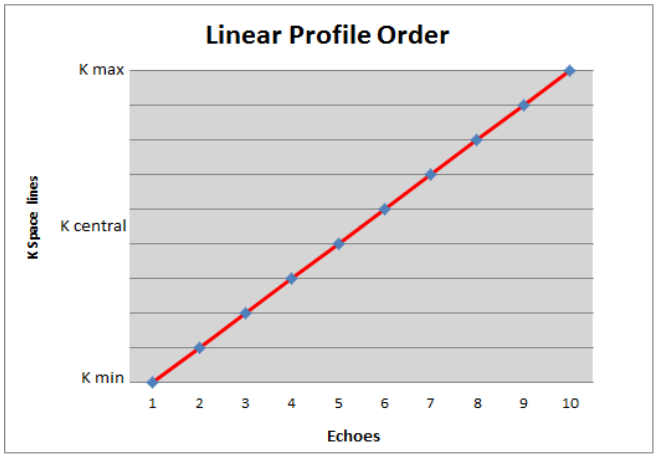


Figure 24: Linear Profile Order

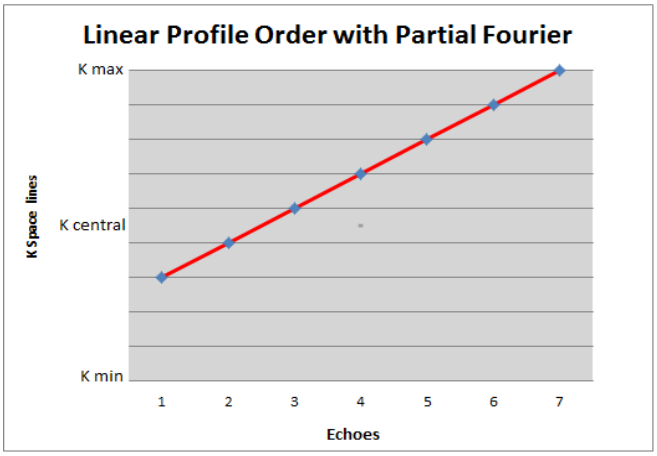


Figure 25: Linear Profile Order with Partial Fourier

9.2.7.2 Reverse Linear Profile Order

A mirrored version of the Linear Profile Order, where the order is set to be linearly decreasing with the increasing echo value (Figure 26).[5]

It provides the same results as Linear Profile Order, except if Partial Fourier is applied (Figure 27). In this instance the effective TE increases.[5]

Figure 26: Reverse Linear Profile Order

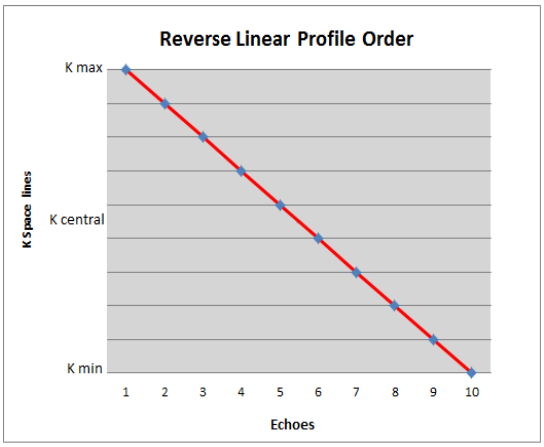
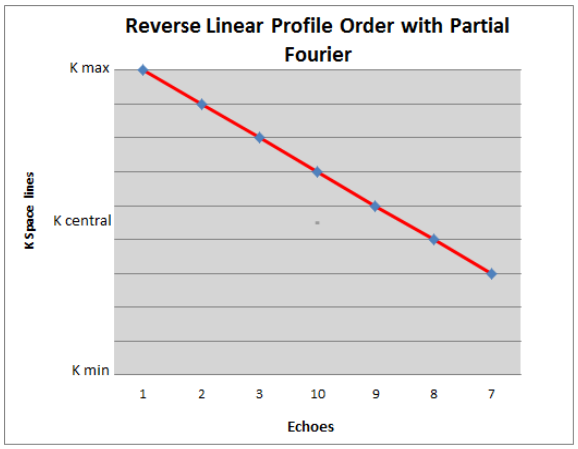


Figure 27: Reverse Linear Profile Order with Partial Fourier



9.2.7.3 Low-High Profile Order

Echoes are ordered into K-Space lines in the following fashion: 0;-1;1;-2;2;-3;3;-4;4;…(Figure 28) Line number zero represents the central line of K-Space.

In a Low-High Profile Order, the Weight is defined by the first echoes, enabling good T1 contrast. Its sensitivity to blurring is quite high, reason why the TF used should be low.

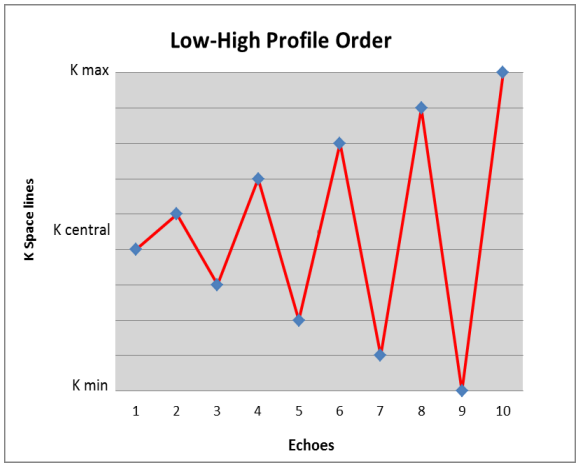


Figure 28: Low-High Profile Order diagram.

9.2.7.4 Asymmetric Profile Order

During sampling, echoes are ordered into the K-Space lines in a pseudo-randomized fashion. The pseudo-randomization is done in a way that ensures the effective TE is equal or similar to the TE defined by the operator. This is particularly true for short and medium TEs.

When this Profile Order is chosen, ES will become independent from the selected TE. This property of Asymmetric Profile Order originates as a necessity from the randomization process and makes the Profile Order very flexible for manipulating Chemical Shift and ET.

Its sensitivity to blurring is higher than any Linear Profile Orders, but both short and intermediate TFs are still acceptable.

Flow artefacts are inherently attenuated with this Profile Order due to the non-sequential (non-organized) filling of lines, yet scanning areas that have intense flow will still generate severe artefact. For this reason, application of flow attenuation techniques is still necessary.

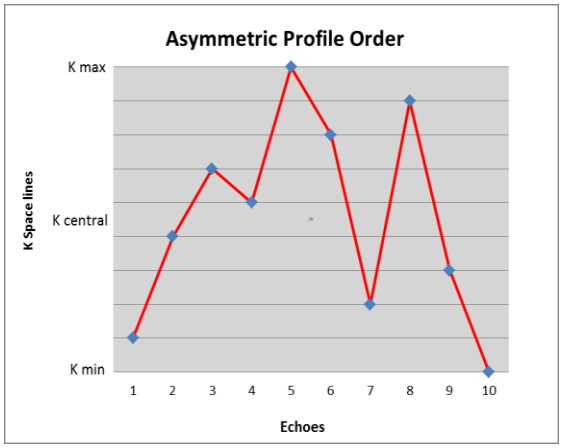


Figure 29: Asymmetric Profile Order diagram

The greatest drawback of Asymmetric Profile Order is its incompatibility with Partial Fourier technique (see subchapter: 11.1.1). Like in Radial acquisition, the non-organized order used to fill K-Space, with Asymmetric Profile Order, makes it impossible for the system to use K-Space's symmetry principles to correct the lack of information introduced by Partial Fourier to reduce TA.

Overall, Asymmetric Profile Order is the preferable choice for most MSK clinical cases mainly because it's more adaptable to manipulation of other parameters; it allows for fairly high TF values; makes Chemical Shift and ET easier to manipulate (because of the non-dependency between ES and TE/TF); flow artefact is partially attenuated; usually provides a better relation between SNR, TA and blurring in the N_{PE}.

In some occasions were both SNR and TA are too high and Sensitivity Encoding (SENSE) factor is already significantly high or can't/shouldn't be employed, it's preferable to change to another Profile Order (according to the desired Weight) in order take advantage of Partial Fourier.

9.2.8 Bandwidth

Bandwidth (BW) in MRI refers to the range of frequencies used by the scanner to either excite protons or sample their relaxation signal. It's used in conjunction with the gradients at two distinct moments of the whole process. Therefore BW is divided into two types: Transmission and Receiving BW.

9.2.8.1 Transmission Bandwidth

The first application of BW, referred as Transmission BW, consists on the range of frequencies used to excite atoms in a discrete slice thickness or slab. It is therefore used when the slice gradient is turned on. The range of the Transmission BW tends to be only of a few kilohertz and can be partially adjusted in some scanners [15].

Both Siemens and Philips scanners allow for direct manipulation of Transmission BW. Three distinct modes for both Transmission BW and gradients are usually available.

The modes available present both advantages and disadvantages. They should be chosen carefully, according to the situation at hand and their correlation with the gradient mode:

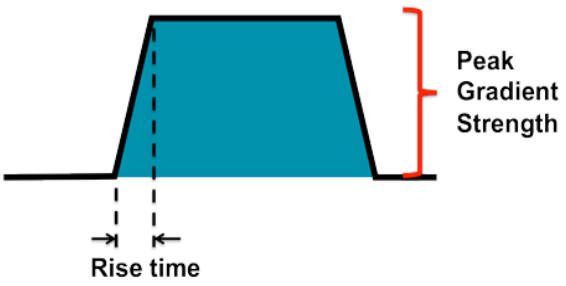


Figure 30: Factors that define a gradient's slew rate. Extracted from [4]

1. Low SAR mode: [15]
    a. A long 3.84ms[3] duration pulse, that allows for a good slice profile;
    b. Reduced risk of crossed talk between slices by reducing the tolerance of the acceptable gap. It allows narrower gaps between slices and provides robust sequential slice acquisition;
    c. Requires longer minimum TEs and TRs due to duration of transmission pulse;
    d. SAR values are reduced due to lower amplitude of pulse.
2. Normal mode: [15]
    a. Balanced mode;
    b. 2.56ms[1] duration pulse with good slice profile;
    c. Optimized SAR behaviour;
    d. Allows for acquisition of long duration scans without breaching SAR limits in the vast majority of the situations.
3. Fast Mode:[15]
    a. Short, 1.28ms[4], duration pulse with compromised slice profile, allowing for possible cross-talk between slices (this can be compensated by opting for interleaved slice acquisition, which will increase the physical gap between slices acquired in the same TR, without having to increase the Slice Gap);
    b. Allows for shorter TEs, TRs and ES, which are fundamental for quick fast and very fast sequences (single-shot, EPI, Ultrafast GE, etc.)
    c. Higher amplitudes used for the Transmission pulse, which leads to increase in SAR;
    d. Fewer susceptibility artefacts, making it an important mode when scanning object with metal implants (see subchapter 14.3 and Chapter 0).

Associated with Transmission BW modes are the Gradient modes, which are respectively: Whisper, Normal and Fast, for Siemens's scanners [15]. For Philips the modes will be respectively: Regular, Default and Maximum.

All three modes affect the Slew Rate of the gradients (ratio between maximal amplitude of the gradient and the time it takes the gradient to go from 0 to its maximum amplitude (Figure 30)).[4]

[3] Value based on a 1,5T Siemens Symphony with TIM Technology.
[4] Value based on a 1,5T Siemens Symphony with TIM Technology.

Increasing the Slew Rate (by choosing Fast mode), will reduce TA by slightly reducing the minimum TE and minimum TR. This will also lead to an increase in noise level and higher potential risk for occurrence of peripheral stimulation. The opposite will occur by selection of Whisper mode [15]

$$Slew\ Rate = \frac{Max.Amplitude}{Rise\ time}$$

Equation 7: simplified calculation of a Gradient's Slew Rate.

9.2.8.2 Receiving Bandwidth

The Receiving BW refers to the range of frequencies used when sampling the signal from an echo. When the echo appears, the frequency (readout) gradient is turned on, and the signal is sampled (registering the signal's amplitude at that moment) through an Analogue-Digital-Converter.[4, 5, 15]

When the gradient is turned on, a one dimensional straight ramp-like row of columns is created. Each column has a slightly different frequency value than the surrounding columns. The middle of the ramp has a frequency value equal to the frequency of the hydrogen atoms exposed to the B_0, while each side of the ramp has a frequency range equal to BW/2 [7, 11, 15]. On the left side of the ramp, frequency values are inferior the frequency at B_0 and on the right side of the ramp frequency values are superior to the frequency at B_0.

Sampling is a time discrete (non-continuous) process. Each sample takes a specific amount of time, defined as Dwell Time, with a length in the order of microseconds. During each Dwell Time, only one measurement is performed.[11, 15]

In normal conditions, one measurement must be done per each column of the row created by the readout gradient (Base Matrix). Therefore the total sampling time of the echo is: [11, 15]

$$Total\ sampling\ time = Dwell\ time * Base\ Matrix$$

Equation 8: Total sampling time of and echo

The period of the sampled signal (electromagnetic wave) is equal to the Dwell Time of the sampling process. It then follows, that the signal's frequency can be calculated through:[15]

$$frequency = \frac{1}{Dwell\ Time}$$

Equation 9: frequency of a sampled MRI signal

The BW of the sequence can be obtained in same fashion using instead the Total sampling time:[15]

$$BW = \frac{1}{Total\ sampling\ time}$$

Equation 10: Receiving bandwidth of a MRI sequence

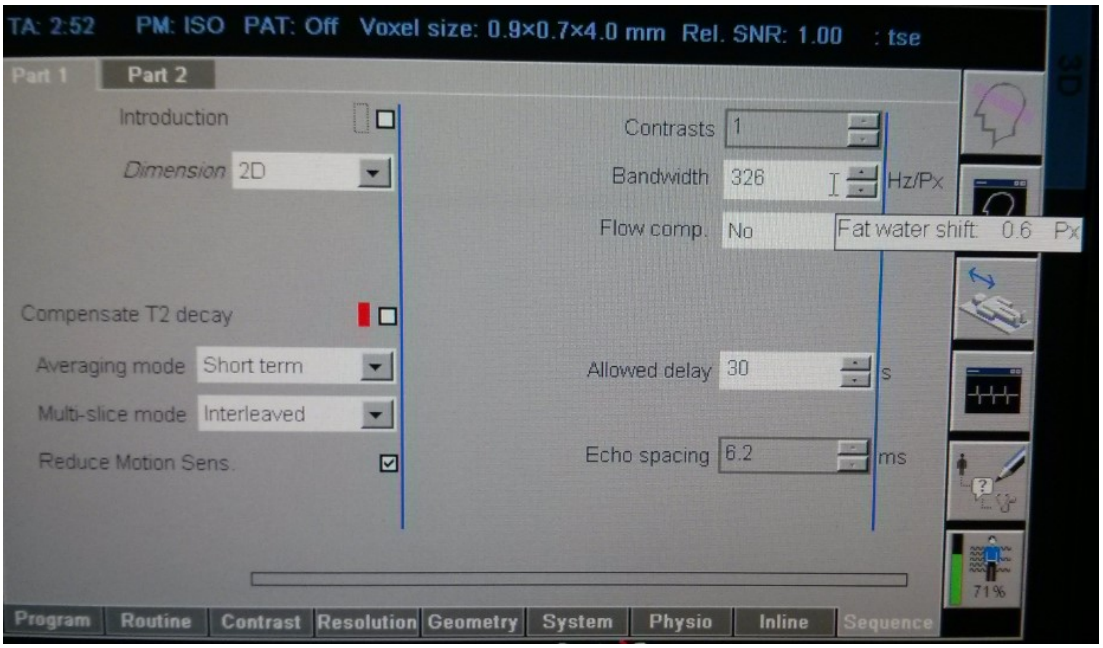


Figure 31: Siemens' Syngo MRI Software. Sequence menu. Receiving BW and Chemical Shift per Pixel.

The unit of the Receiving BW is Hz/pixel (Figure 31) and exposes, in a simplified way, the range of frequencies sampled within one pixel of the image.[7, 11, 15]

Why is the range of frequencies per pixel important? The simplified answer is: Aliasing and SNR.

As exposed by the Nyquist Theorem (see subchapter 6.5.5) the sampling frequency will dictate how similar the sampled wave and the real wave are. The higher the BW, the more similar both waves will be and the less likely will be for Aliasing or Susceptibility artefacts to appear.

The relation between BW (both Transmission and Receiving) and SNR is a cornerstone of sequence manipulation (see subchapters: 6.5.5 & 14.1 and Chapters 20 & 0). Mathematically is possible to see that SNR decreases in a hyperbolic fashion as the Receiving BW increases (Figure 78). It's a known fact in Science that, whenever large samples of data are considered, precision and accuracy often conspire against each other. This is exactly what occurs in the relation between SNR and BW. Minor variations naturally occur within the induced wave due to interference of the Scanner's and Coil's electronics. These variations are not representative of the true electromagnetic wave produced by atoms and so they represent false information. This is referred as Noise and can have several patterns of statistical distribution (e.g. Poisson distribution), depending on the cause of the Noise. As BW gets higher, the number of possible frequency values increase and the sampling rate improves. Under these conditions, it becomes easier for the system to register those minor variations and so the likeliness that Noise will be sampled improves. The likelihood of sampling True Signal remains constant with changes to BW, since it only depends on the number of samples performed. It then follows that because the probability for sampling Noise increases but the probability for sampling True Signal remains constant, then SNR gets worse at higher BWs.

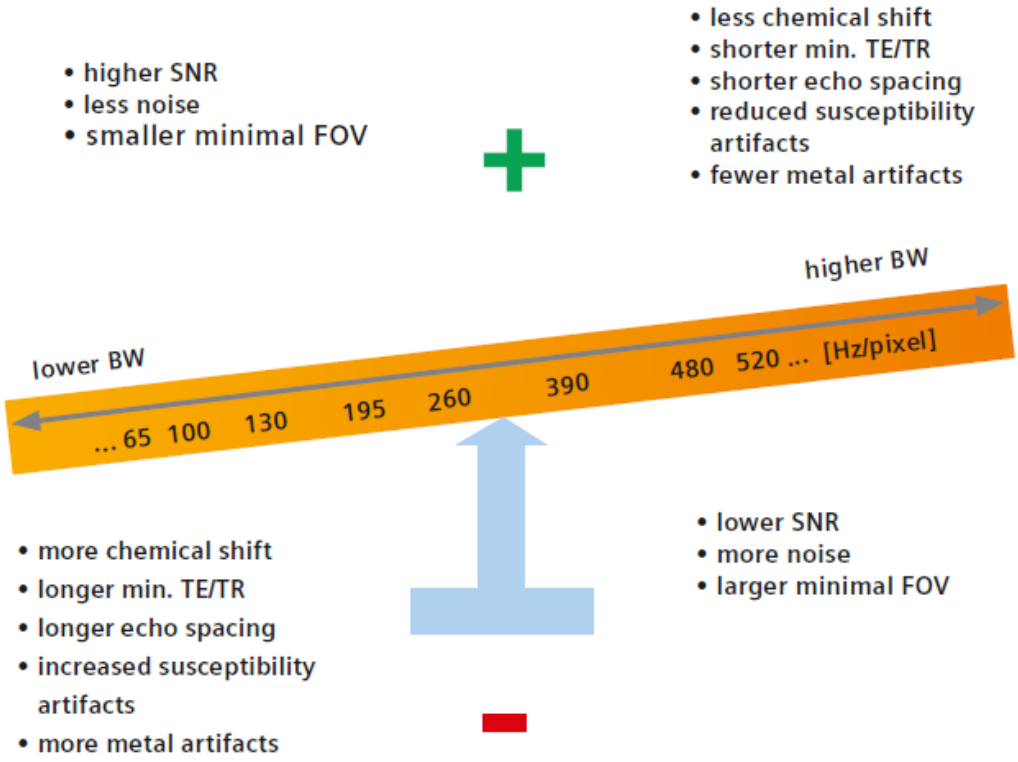


Figure 32: Relation between Receiving Bandwidth and other MRI parameters. Extracted from [15]

As a MRI parameter, BW is a core aspect to take into account when manipulating a sequence. It defines the rate at which the signal from the echo is sampled; directly affects SNR, minimum TR, TE, ES and FOV, total sampling time and the severity of occurrence of important artefacts, particularly Chemical Shift, Susceptibility and in-plane Aliasing.

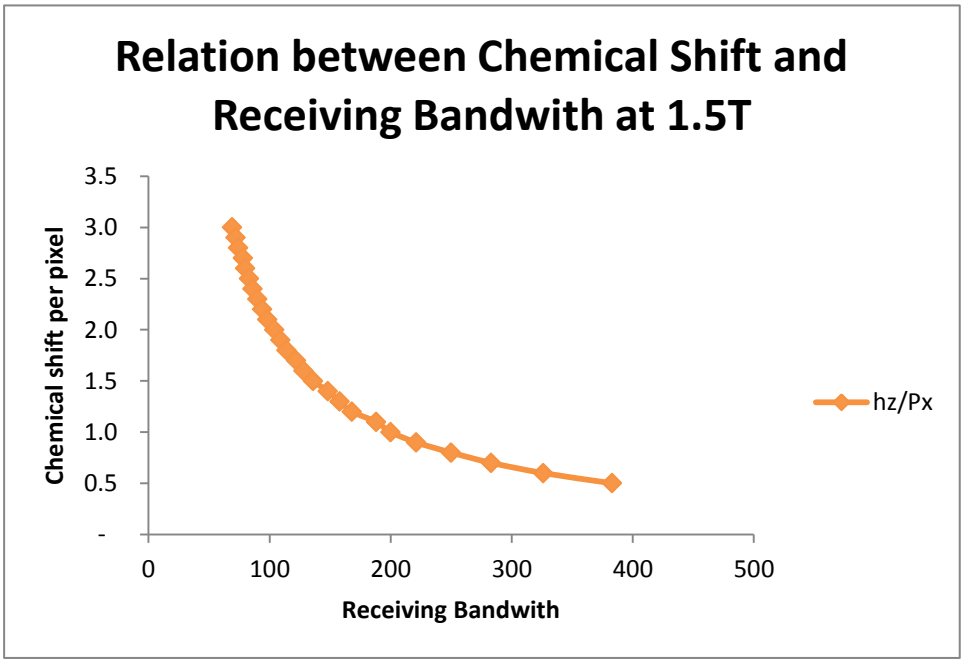


Figure 33: Relation between Chemical Shift and Receiving Bandwidth at 1.5T. Data collected from a Siemens Symphony.

9.2.9 Motion smoothing

Slight modifications are made randomly to the ET's echoes, in TSE, without changing the profile order or the effective TE. The random modifications produce some reduction/attenuation of repetitive artefacts (breathing, ghosting, etc.). It is only effective when there is a large number of echoes.[5]

Motion smoothing does not interact will other MR parameters.

9.2.10 Flow compensation

It partially attenuates in-plane Flow Artefacts (see subchapter 14.2.2.1.3) caused by fluid motion (e.g. Blood flow) along the acquired plane. It works by correcting magnitude variations produced in-plane, within each pixel, by proton phase dispersion. Magnitude correction is performed by compensating the dephasing part of the read-out gradient, between the excitation pulse and the refocusing pulse, with the first half of the read-out gradient, after the refocusing pulse.[5] This compensation balances the phase between stationary and moving protons, resulting in attenuation of Flow Artefacts on the phase direction only.

Its effectiveness is limited and varies substantially from scan to scan because flow artefacts are dependent on many factors, including FOV, plane orientation and the patient's heart rate.

It can also be applied in the slice direction in order to correct through plane flow artefacts. Using saturation bands positioned adjacent to the FOV where the blood/fluid flows into the FOV can also correct this problem. This way of flow compensation using sat bands tends to be more effective in MSK imaging that using Flow Compensation.

Usually, better Flow Compensation is achieved in GE sequences (due to absence of in and out-flow effects) and with larger values for the majority of parameters: FOV, TE, ES, SNR, etc.

Interaction between Flow Compensation and other sequence parameters vary depending on the sequence used:

- In SE, IR and MIX (combination of Multi-echo TSE and TIR sequences) Flow Compensation as no influence in other parameters, apart from increase in the minimum TR, but compensation may not occur
- In TSE sequences, SNR and Chemical Shift per pixel will often be reduced with increased TA (due to SNR adjustment and/or increase in the minimum TR). Full compensation is possible when ES is short enough;
- In GE sequences, tuning on Flow Compensation may lead to increased TE and reduction of SNR and Chemical Shift.

9.2.11 Slice Turbo Factor

Parameter only available for 3D sequences. It defines the number of slices acquired per ET [28]. It affects the length of the ET in a direct linear proportion and TA in an inverse linear proportion. As explained earlier, blurring effects are substantially increased with the increase of ET's length.

9.2.12 Image Normalization

It's is used to reduce significant brightness variations throughout the FOV and attenuates Shading Artefacts. Brightness variations can difficult visualization of areas far away from, or too close to, the receiving coil. It's known in Philips' Scanners as CLEAR and in Siemens' scanner as Normalize or Pre-Scan Normalize. When used, the image's brightness is averaged in the entire FOV based on the sensitivity and position of the coils. This information is obtained in Philips' scanners by a Reference Scan (acquired previous to the sequence(s)) and on Siemens's scanners prior to every sequence acquisition (for Pre-Scan Normalize only. On the other hand, Normalize relies in adjusting the image histogram post image reconstruction).

This parameter should be turned on in almost all situations, since it doesn't conflict with any parameter and improves the overall visual appearance of the images without compromising the image's Weight or spatial resolution.

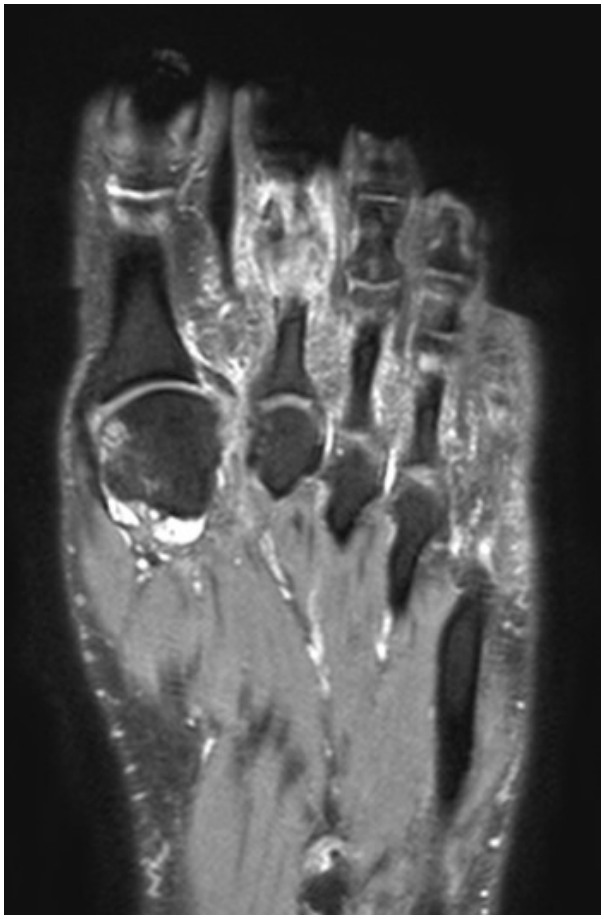


Figure 34: Axial PD_FS TSE of the left foot. Loss of signal at the tip of the toes as a result of using Image Normalization.

In some peculiar instances (Figure 34), where either Susceptibility-related Artefacts or J-decoupling are quite significant, Image Normalization generates, respectively, areas of extreme hypo intensity or extreme hyper intensity. It best seen as non-grainy signal drop-out at the tip of the extremities (fingers and toes. see Figure 34) in Fat saturated sequences. In such instances, repeating the sequence without Image Normalization or with other method for brightness balance is advisable.

9.2.13 Scan Percentage

An unusual parameter that is only available in scanners by certain manufacturers (e.g. Philips). It's not defined by the operator, but is instead automatically adjusted by the scanner to the closest allowable value based on the operator's input for the TF [5] and some K-Space manipulating parameters (e.g. Half Fourier)

It's designed to reduce TA at the cost of image sharpness, by reducing (percentagewise) the number of K-Space points sampled within each encoded line. For the first 25% sampling reduction is

done by applying a, gradually increasing, K-Space shutter. After that, the TA reduction is done at the cost of in-plane spatial resolution in the Enc. Dir.[5]

Scan Percentage should be kept within 80% and 100% to assure decent image quality. For values below 80%, Aliasing artefact will likely become too severe. [5]

This unusual trade-off is of small interest for MSK imaging, where high resolution and very sharp images are very desirable technical aspects. Nevertheless, it is important be bear it in mind when manipulating parameters and understand the potentials effects of the value that Scan Percentage is set as at the moment of acquisition. Scan Percentage is frequently overlooked as the potential cause for bad image quality, because the operator is unaware of what the values for this parameter represent.

There is no loss of SNR with decreasing Scan Percentage values [5], making this parameter interesting to certain types of sequences/images that use low resolution and/or intrinsically have very low SNR (e.g. Spectroscopy, Diffusion Weighted Imaging (DWI)).

9.2.14 Over-Continuous slices

For sequences that used volumetric acquisition (3D) (see subchapter 11.1.7), additional parameters are available in some scanners. One of the most important ones is called: Over-Continuous (Philips' nomenclature). This parameter doubles the defined thickness of the acquired voxel and then interpolates the voxel's size into the defined thickness during the reconstruction process.

Doubling the acquired voxel's doubles the SNR and halves the TA. Such trade-off is rare and very desirable in MRI, particularly because the drawback of lower resolution is mitigated by the interpolation aspect of this parameter. Image sharpness in the Z axis will likely deteriorate as a result of interpolating the signal intensity of one voxel into two. If ET length is already large, then Over-Continuous slices should be used because the sharpness deterioration in non-significant, compared with the blurriness from the long ET.

10 Tissue saturation

Within MRI's potential to generate multiple contrasts, tissue saturation is probably the most important ability for clinical use.

Their use in MSK is fundamental to clarify doubts that would, otherwise, be a nightmare in diagnostics. Characterization of tissues is mostly dependent on their signal intensity but some very different tissues present similar intensity in one or multiple types of MRI Weights. This is particularly true for fat tissue in MSK. Being able to distinguish between fat and other high signal tissues, or be able to remove the glare effect that fat's signal can cause in its vicinity, is fundamental for a correct diagnosis.

So the question isn't: "Is tissue saturation useful?" The question is: "How can we properly saturate a specific tissue?"

10.1 Short Tau Inversion Recovery (STIR)

Inversion Recovery sequences (see subchapter 8.2.2) were the first form of tissue saturation developed in MRI. They were conceived in response to two important necessities: the need to saturate fat tissue in MSK and abdomen-pelvic examinations and the need to saturate CSF in Neurological examinations.

The principle is based around the application of a 180º RF inversion pulse, before the 90º RF excitation pulse at a specific time, considering the T1 of the tissue that needs to be saturated. After the inversion pulse the M_L is negative and starts varying towards its original positive value. ML will cross the null value at some point in time, and so there is a specific temporal interval between the moment the inversion pulse is applied and the moment each tissue possesses a null M_L. This temporal interval is called Time of Inversion (TI) or Tau (because its physical symbol is τ) (Figure 35). Just like T1, it's specific for every tissue ($0.69*T1_{tissue}$) and dependent on B_0 (since T1 is (

Table 1)).[7, 11, 30]

Full saturation of a specific tissue can be achieved by carefully defining that the time interval between the inversion RF pulse and the excitation RF pulse is equal to the tissue's TI. In the case of fatty tissue, TI is around 150ms at 1.5T and 190ms at 3T. Choosing a TI close to the tissue's TI will still generate saturation, although only partially.

Saturation is obtained because, when the excitation pulse is applied, at the moment that the M_L of a specific tissue is null, the protons from such tissue are insensitive to the RF excitation and won't abide by the principles of precession resonance.

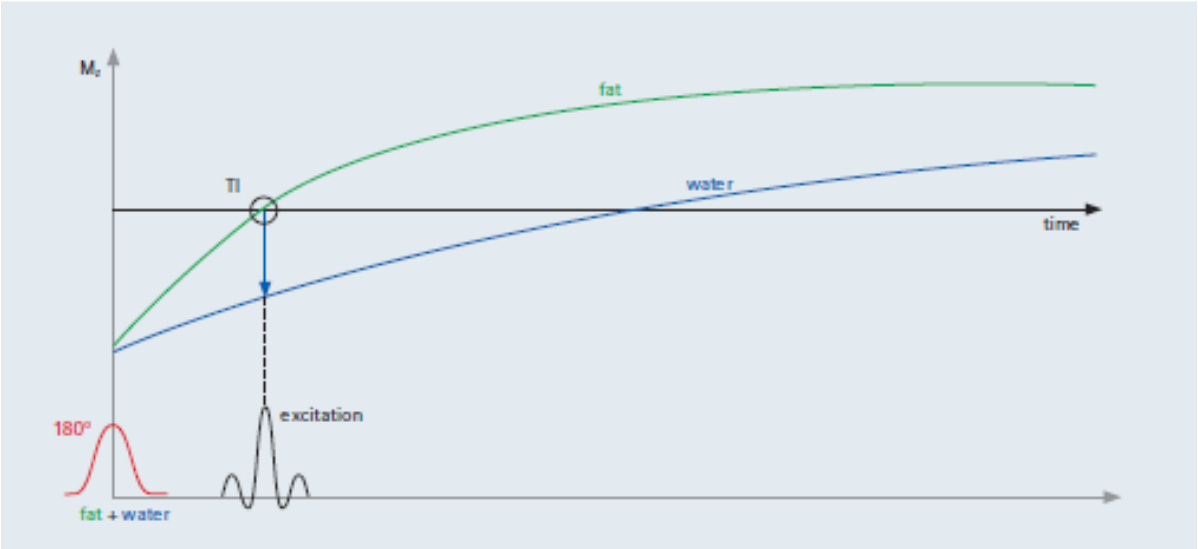


Figure 35: IR technique for fat tissue saturation. [30]

The images produced have inverted T1 contrast (which is very similar to T2 with fat saturation), because, at the moment of excitation, the negative signal from tissues that aren't been saturated is flipped back to their mirror positive value. STIR sequences are quite robust to B_0 inhomogeneity and to susceptibility artefacts (compared with other tissue saturation methods) yet susceptibility-related artefacts will appear at the edges of the FOV, when large FOVs are used.

STIR or any kind of IR sequence should not be used in conjunction with MRI contrast agents, because the contrast agents shorten the tissues' T1, meaning that the TI of any tissue will also change and produce improper saturation of the tissue intended on being saturated. Saturation of tissues that shouldn't be saturated may also occur.[11] The final result is potential distortion of the images' diagnostic information.

10.2 Spectral Fat Saturation (Fat Sat)

Unfortunately IR methods affect the tissue's contrast, creating images with a Weight that differs from T1, PD and T2 [20, 30, 31]. This makes STIR images difficult to compare with T1 images, especially when contrast enhanced images are acquired. To solve this problem, Fat Sat (Siemens's commercial name. See page 139 for other manufacturers) is currently used in almost all MSK cases, particularly in cases that require injection of contrast agents.

Fat Sat works by taking advantage of the chemical shift between Water and Fat [30] (see subchapters 9.2.8.2, 14.2.1.2 & 14.3) in order to differentiate hydrogen in Fat from hydrogen in Water. Saturation is done by selectively exciting fat-bound hydrogen, using a very narrow RF excitation pulse (with a frequency value equal to the precession frequency of fat-bound hydrogen), followed by the application of a spoiling gradient that partly or almost completely nullifies the relaxation process of fat tissue (destroying the signal produced by fat-bound hydrogen (Figure 36)). This is done prior to the application of the normal RF excitation pulse (during the TR's pre-pulse phase) and causes fat-bound hydrogen to become insensitive (because they are in a state of "saturation" induced by the spoiling) to the RF excitation pulse [30]. This means that the large majority of fat-bound hydrogen doesn't contribute to the signal emitted during precession decay and so fatty tissue appears dark, or at least partly dark, on the reconstructed images.

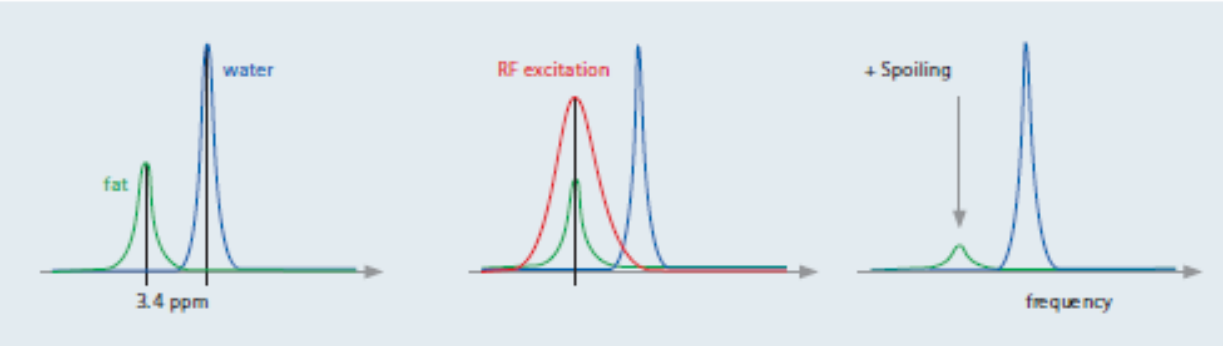


Figure 36: Spectral Fat saturation. Extracted from [30]

With this technique only the precession decay of the fat-bound hydrogen is scrambled. The image contrast of the water-bound hydrogen atoms is maintained exactly as a regular T1, PD or T2, facilitating comparison with non-fat saturated images or in contrast enhanced examinations.

The drawbacks of this method are: a slight increase in the TR (due to the need to saturate fat before applying the main RF excitation pulse) [30]; significant reduction in SNR, due to the lack of contribution of fat-bound hydrogen to the overall signal; high dependency on B_0 homogeneity and

gradients [11, 20, 30, 31], since the saturation process relies on selective excitation of specific frequencies and in the chemical shift on the area where the pulse is applied.

In practice, good Fat saturation is hard or impossible to achieve at low field strengths (<1T), since this technique requires good separation between Water and Fat resonance peaks. [36]

Adequate fat saturation is almost certain to fail if metal implants are present, or a large FOV is used or if the body part scanned is either oddly shaped or too close to the bore's wall/tunnel entrances, due to the this technique's high B_0 homogeneity dependency.[11, 30, 31] In situations like the ones mentioned or when FAT SAT fails for unknown reasons, other fat saturation techniques should be applied.

Additionally, FAT SAT can be applied in two modes: weak or strong. In the first one fat tissue is only partially saturated (appearing dark grey to light grey on images) while in the second one fat is almost totally saturated (appearing black to dark grey on images).[30]

10.3 Spectral selective Inversion Recovery (SPIR) and Spectral selective Adiabatic Inversion Recovery (SPAIR)

Being a sort of mix between Fat Sat and IR, SPIR/SPAIR (SPIR is available in Philips' scanners and SPAIR in both Philips and Siemens) present themselves as more robust techniques than Fat Sat for saturation of fat tissue. Like with Fat Sat the image contrast doesn't change because only fat-bound hydrogen is disturbed during pre-pulse saturation. Detailed contrast is slightly reduced, but tissue saturation is more complete and homogeneous throughout the entire image, than with Fat Sat (Figure 37). SPIR/SPAIR is also more resistant the B_0 inhomogeneity, particularly when scanning objects close to the bore's wall.

Although not as resistant as STIR sequences, particularly when metal implants are present, SPIR/SPAIR may prevent fluctuations in tissue saturation. I these instances it will produce areas of extreme high or low signal at the edges of the FOV, where fat saturation is not performed properly (Figure 38).[5, 20, 30, 36]

The minimal TR is significantly higher than on sequences using Fat Sat.[20, 30, 36] This is especially true when a low TF is used.

SPIR works by application of an inversion pulse during the pre-pulse phase, in the same fashion as a STIR sequence. The difference relies on the type of inversion pulse used. Instead of a generic inversion pulse that affects all tissues, SPIR uses an inverted version of the fat-selective pulse used in Fat Sat [5, 7, 11, 26, 36], which allows for selective inversion of fat tissue and then applies a signal spoiling gradient (to destroy the signal from fat tissue's M_T vector). This guarantees that the image's contrast is not inverted (as in a STIR image) while at the same time providing full and robust fat saturation. [36]

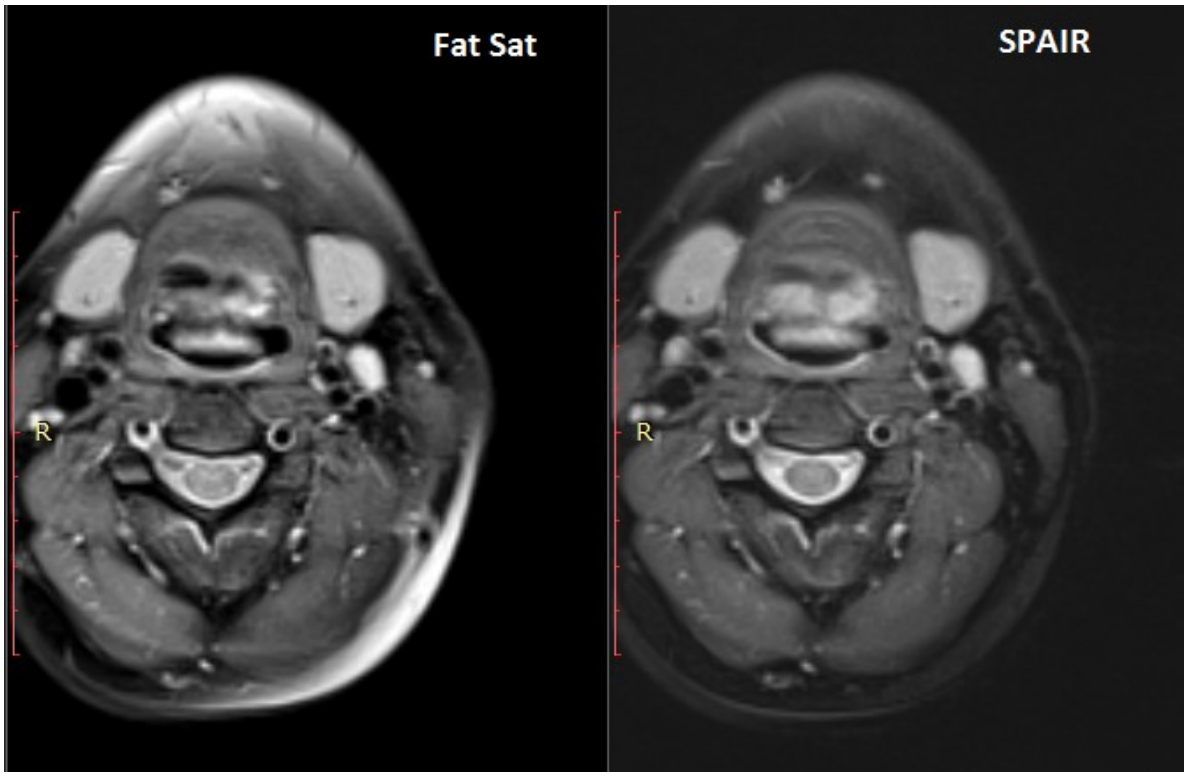


Figure 37: Axial Proton Density with Fat Saturation (PD_FS) of the neck with Fat Sat and SPAIR respectively. Improper saturation of fatty tissue is seen on images acquired with Fat Sat as a result of the complex shape of neck area. Images with SPAIR do not present this problem, but a slight loss in detailed contrast within the spinal cord.

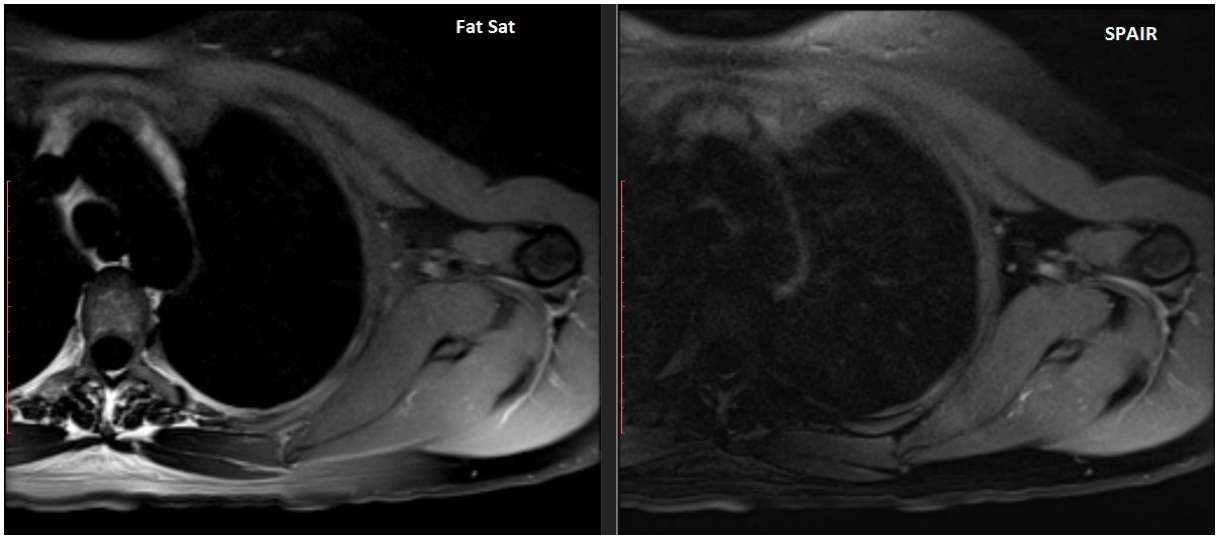


Figure 38: Axial PD_FS acquired with Fat Sat and SPAIR respectively. Different results are obtained when fat saturation fails. The excitation pulse in Fat Sat enhances the signal moderately in areas where the technique fails (around spine and deltoid). SPAIR on the other hand produces areas of very low signal (spine) due to the use of an inversion pulse (the exact opposite may also occur with SPAIR). There are more are of improper fat saturation with Fat Sat than with SPAIR. Additionally, reduced SNR is seen in SPAIR images compared with Fat Sat images.

SPAIR works much like SPIR, but the selective inversion pulse applied is adiabatic, which provides an improved independency between the saturation and B_0 inhomogeneity. [5, 20, 30, 36] In practice, the adiabatic pulse is not as effective as one would hope.

Nevertheless, SPIR and SPAIR require good discrimination between water and fat resonance peaks. They can be applied to T1, PD or T2 weighted images [36] and can be applied in two modes: weak/strong (for Siemens' scanners) or 1/2 (for Philips' Scanners).[5, 36]

As suggested earlier (Figure 38), SPIR and SPAIR require a bit more SNR than Fat Sat as a result of using an inversion pulse.

These techniques also slightly degrade detailed contrast (Figure 37). To attenuate this, Philips [5] recommends that SPIR should be used for T1 and PD images, and SPAIR for T2 images.

10.4 DIXON technique

Used more and more nowadays (especially in abdomen examinations), the DIXON technique is an interesting method that allows for reconstruction of up to four types of different image weights at the same time: In-phase, Out-phase, Water only, and Fat only images. It relies on the chemical shift between Water and Fat to discriminate the signal from the two different tissues, during acquisition. The signal is sampled at two distinct moments (defined by two TEs. Table 8): In the first moment water-bound and fat-bound protons are precessing in the same phase and in the second moment when they are precessing in opposing phase.[5, 11, 30, 31, 36]

The result of the signal discrimination are images with two weights (in-phase and out-phase) that can then be used to generate water-only images and fat-only images.[5, 11, 30, 31, 36]

Table 8: Reference TE values for In-Phase and Out-of-Phase, between water and fat tissue. Extracted from [5]

	0.5T	1.0T	1.5T	3.0T
in phase TE [ms]	13.8 27.6	6.9 13.8 20.7 27.6	4.6 9.2 13.8 18.4 23.0	2.3 4.6 6.9 9.2 11.5
Water and fat are in phase when the TE is a multiple of ... [ms]	13.8	6.9	4.6	2.3
out of phase TE [ms]	6.9 20.7	3.45 10.35 17.25 24.15	2.3 6.9 11.5 16.1 20.7	1.15 3.45 5.75 8.05 10.35

DIXON is achievable in both GE and SE sequences, but is harder to obtain in SE sequences because of the presence of refocusing pulses that realign all protons in the same phase.[5, 11, 36] It possesses good SNR, since all tissues produce signal that is sampled, and has more resistance to B_0 inhomogeneities than spectral saturation techniques (Fat Sat, SPIR and SPAIR). [5, 11, 30, 31, 36]

The effective TA for DIXON is long (on average around 5 to 6 min for GE DIXON. It's longer for SE DIXON), but can be seen as short if considering that it allows for simultaneous acquisition of multiple types of images.[36]

10.5 Water Excitation

Water Excitation (Figure 39) is a form of fat saturation that relies on enhancing the signal from water-bound hydrogen over the signal coming from fat-bound hydrogen.

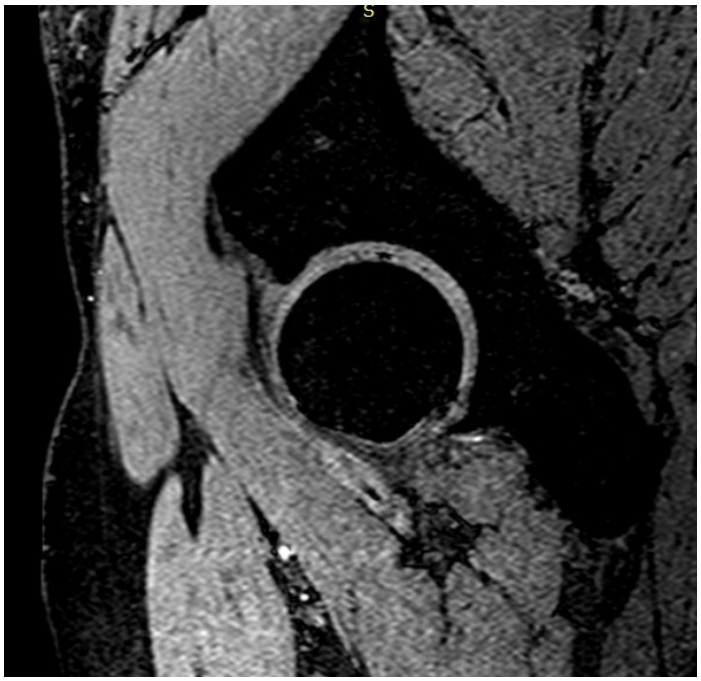


Figure 39: Water Excitation Sagittal plane of hip.

Applying the physical principles of wave destruction/construction, a modelling complex binomial wave is applied over the normal signal spectrum. The modelling wave's function is set so that it's constructive (in-phase) for the water resonance peak and destructive (out-phase) for fat's peak. [30, 36]

This method for enhancing Water's signal negates the need for a normal 90º RF excitation pulse, because the complex wave forces the water-bound hydrogen to flip 90º. Fat-bound hydrogen is also flipped by the complex wave, but because the pulses are not in-phase with them, the resulting flip angle obtained for fat-bound hydrogen is equal to 0. [36]

The complex wave in composed by elemental waves with distinct flip angles. The flip angle of each elemental is not important as long as the temporal delays between elemental waves are set properly so that the resulting flip angle for Fat equals 0 degrees. [36]

The result is an image with very high contrast between fluids/cartilage and fat (Figure 39). Contrast wise is similar to T1 with fat saturation but with higher structural detail and can only be produced by GE sequences. Its main clinical use is cartilage measurement and segmentation.

11 Acquisition methods

11.1 K-Space filling

To obtain an image from the sampled signal, it is fundamental to encode it in a way that allows us to localize the source of the signal in space, during sampling. As discussed previously, this is done by using an imaginary planar grid, defined as K-Space, which can be either 2D or 3D. K-Space is the realm of frequencies and possesses some interesting characteristics, being the most important of all its spatial symmetry in relation to the central point of the grid.[7, 8, 11, 13]

K-Space is also known in the Tomography field as Fourier's Space. This is because if the 2D Fourier Transform is applied to an image, a representation of the image in its elemental frequencies will be produced. [3, 8, 11, 13] This is exactly what K-Space is.

The Fourier Transform (Equation 12) is, simply put; a mathematical algorithm that decomposes a complex signal/wave into simple periodic functions (sinusoids) and in the process converts something that exists in Real Space (R Space) into the imaginary space of frequencies. [2, 3, 7, 8, 11, 13]

These sinusoids are much more easily processed mathematically than a standard image in R Space. This is the reason why such algorithms are used for image reconstruction. [2, 3, 7, 8, 11, 13]

The Fourier Transform is just one of many methods (e.g. Radon Transform) to mathematically decompose an image or signal. It's "simplicity" and how well it adjusts to the shape and way how the K-Space is normally filled in MRI, makes the Fourier Transform the preferred method for image reconstruction. [2, 3, 13] To speed up the reconstruction process, normally a discrete form of the Fourier Transform, called Fast Fourier Transform is used (Equation 11): [2, 3, 13]

$$F(x,y) = \int_{-\infty}^{\infty}\int_{-\infty}^{\infty} f(k_x,k_y)\, e^{-2\pi i (k_x x + k_y y)}\, dk_x\, dk_y$$
$$f(k_x,k_y) = \int_{-\infty}^{\infty}\int_{-\infty}^{\infty} F(x,y)\, e^{2\pi i (k_x x + k_y y)}\, dx\, dy.$$

Equation 11: Fast Fourier Transform. Simplified form of Fourier Transform for speed up computation processes.[2]

$$\sum_{n=0}^{N-1} a_n\, e^{-2\pi i n k/N} = \sum_{n=0}^{N/2-1} a_{2n}\, e^{-2\pi i (2n) k/N}$$
$$+ \sum_{n=0}^{N/2-1} a_{2n+1}\, e^{-2\pi i (2n+1) k/N}$$
$$= \sum_{n=0}^{N/2-1} a_n^{\text{even}}\, e^{-2\pi i n k/(N/2)}$$
$$+ e^{-2\pi i k/N} \sum_{n=0}^{N/2-1} a_n^{\text{odd}}\, e^{-2\pi i n k/(N/2)},$$

Equation 12: 2D Fourier Transform.[2]

In MRI, we start from sampled signals which are then used to reconstruct the visual image. This means that we go from K-Space to R Space, which is the opposite way Fourier Transform works. To solve this problem an inversion of Fourier Transform is applied to the filled K-Space.[3, 7, 8, 11]

$$\begin{aligned}\mathcal{F}_x\left(-PV\frac{1}{\pi x}\right)(k) &= -\frac{1}{\pi}PV\int_{-\infty}^{\infty}\frac{e^{-2\pi i k x}}{x}dx \\ &= PV\int_{-\infty}^{\infty}\frac{\cos(2\pi k x)-i\sin(2\pi k x)}{x}dx \\ &= \begin{cases}-\frac{2i}{\pi}\int_{0}^{\infty}\frac{\sin(2\pi k x)}{x}dx & \text{for } k<0 \\ \frac{2i}{\pi}\int_{0}^{\infty}\frac{\sin(2\pi k x)}{x}dx & \text{for } k>0\end{cases} \\ &= \begin{cases}-i & \text{for } k<0 \\ i & \text{for } k>0,\end{cases}\end{aligned}$$

Equation 13: Inverted Fourier Transform.[2]

11.1.1 Partial/Half Fourier

As discussed above, both K-Space and Fourier Space are, in practice, the same thing. One important aspect of this space is its spatial symmetry.[3, 7, 8, 11, 13] Unlike R Space, K-Space is symmetric to itself in any direction that crosses its central point. These means that if an imaginary straight line/plane is traced, crossing the K-Space central point, and divides it into two regions of the same size, then those two regions will be a mirror image of each other, regardless of the direction that the line/plane is traced. This symmetry applies in both 2D and 3D K-Space and can be exploited to reduce the TA of most MRI sequences by reducing the number of N_{PE} acquired. This technique is called Half Fourier or Partial Fourier (also known as Half Scan). [3, 5, 7, 8, 11, 13, 18, 22, 37] It only works if an entire region is filled in no gap between the lines, reason why this technique is incompatible with Asymmetric Profile Order (see subchapter 9.2.7.4).[5]

Theoretically, if half of the information used to fill K-Space is known, then it's possible to fill the rest of K-Space by simply filling the other half with an inverted/flipped version of the known data.[5, 7, 8, 11, 13] It's similar to cutting an apple in half. Both sides of the inside of the apple are exactly the same, just a mirror version of each other. This idea could be explorer further by acquiring only one quarter of K-Space (Half Fourier in conjunction with Partial Echo (see subchapter below)) and then flipping it to obtain one half of K-Space, and then flipping that half to fill the whole space without risk of distorting any information.

However, in practice things are never so simple and not even acquiring half of K-Space guarantees that information won't be distorted.[3, 7, 8, 11] The presence of Noise during the sampled process leads to images with low SNR (grainy images) when Half Fourier is used (because the number of sampled performed is low) and can also generate weird artefacts when in conjunction with Phase miss registration (normally introduced by motion during signal acquisition) (see subchapter 11.2.2).

Half Fourier is still used quite often, particularly in conjunction with EPI [25, 28] (see subchapter 0) acquisition for ultrafast scanning and to reduce SAR.

This parameter is operator dependent and several options are available to be chosen. Normally, at least 60% of K-Space needs to be filled in order to guarantee that no distortion occurs

when Half Fourier is used and that SNR is enough to produce images with acceptable quality in a low TA.[3, 7, 11, 13]

11.1.2 Partial Echo

Partial Echo is quite similar to Half Fourier, but instead of a reduction in acquired N_{PE}, a reduction is the N_{FE} is obtained by not acquiring the first half of each echo (echoes are also symmetric). Acquiring only half of each echo reduces TA because the readout gradient (frequency gradient) does not require having a dephasing lobe and so the length of TR and ET is reduced. [7, 11, 13]

With this technique SNR is also reduced by √2, since only half each echo is sampled. The decrease is partially compensated if the chosen/minimum TE is also decreased.[7, 11, 13]

11.1.3 Cartesian acquisition

The most simple and still most often used acquisition method for filling the K-Space is the Cartesian acquisition.

It consists in acquiring single Phase lines separately (Figure 40), one by one and slightly increasing or decreasing, accordingly, the amplitude of the Phase Gradient between lines.[3, 5, 7, 8, 11, 13, 22, 24]

Cartesian is the slowest method of K-Space filling at it has some sensitivity to motion artefacts, but it possesses many advantages over the other acquisition methods:

- Even distribution of sampled points throughout K-Space;
- Can be performed with small or long ETs (even without K-Space segmentation);
- Most robust acquisition method to susceptibility artefacts and signal miss registration;
- Most robust acquisition method to blurring effects, enabling acquisition of best resolution and image sharpness possible;
- Useful to acquire all types of image type/weight;
- Can be used for all types of sequences;
- Doesn't conflict with the majority of MR parameters (including most Profile Orders), particularly the most commonly used parameters;
- Doesn't require complex and/or ultrafast alternations within or between gradients, leading to lower acoustic noise.

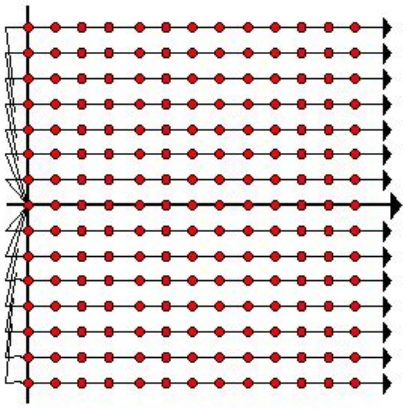


Figure 40: K-Space filling by Cartesian method. Adapted from [8]

11.1.4 Elliptical sampling acquisition

K-Space is sampled in an elliptical fashion, resulting on the edges of the K-Space not being sampled and so not included in the reconstruction data. This allows for a 25% TA reduction without substantial degradation of image quality, since the edges of the K-Space have as small contribute to image resolution and none to image contrast.[5, 8, 22]

Elliptical sampling is not recommended when parallel imaging techniques are being used.[5]

It's an unusual acquisition method, not often available.

11.1.5 Radial acquisition

Radial acquisition refers to a technique to fill the K-Space in a way that the lines of signal acquisition are spatially separated from one another not by their phase value (like in Cartesian acquisition) but by a rotation angle with the centre of the lines being the centre of the K-Space (Figure 41).[3, 5, 7, 8, 11, 22] The angle of rotation between lines is defined by the number of lines acquired in the following fashion: [5]

$$Angle\ of\ rotation = 180º / Number\ of\ encoded\ lines \text{ (or Blades)}$$

Equation 14: angle of rotation on radial acquisition

This method mimics the way non-linear tomography acquires signal, seen in modalities like non-helical CT or 2D PET.

Just like in CT and PET, the FOV is circular and there is an unbalance between the amount of sampled high frequencies (outer area of K-Space) and low frequencies (central area of K-Space) (Figure 41), giving this type of acquisition inherent low band-pass (see subchapter 13.1) properties. This results in images with good contrast but lower sharpness and structural detail (lower resolution).

Small or linear motion artefacts are reduced or nullified (ideally) by two reasons:

1. Information regarding motion and flow artefact is normally stored within the outer areas of K-Space (which is this case are under sampled);
2. Only very few lines will be parallel to the direction of the motion, minimizing the spatial encoding error that motion produces and leads to generating the artefact.

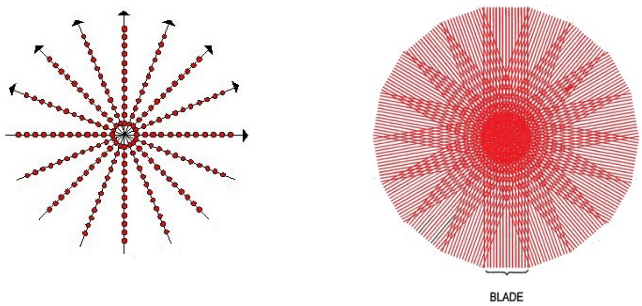


Figure 41: K-Space filling by radial method. Adapted from [8]

Current day radial acquisition (commercial names: BLADE for Siemens and MultiVane for Philips. See page 139) doesn't rely on single encoding lines crossing each other, but instead, blades are used in substitution of single lines. A blade is a group of parallel straight encoding lines.[5] Using blades instead of single lines, partially mitigates the unbalanced sampling between low and high frequencies. It also reduces TA by acquiring multiple encoding lines (the ones composing one blade), of each image, per TR (longer ET).[5, 7, 8, 11, 24]

The order in which the lines within a single blade are acquired follows the Profile Order's pattern chosen (just like Cartesian acquisition), preserving, in general, the concept of effective echo time within TSE.[5]

Radial acquisition, with blades, is only available for use if the following conditions apply:[5]

- Only TSE with a minimum TF (minimum TF is 4 for Philips' scanners);
- Only single echo sequences;
- Only 2D scans;
- Only possible with the following profile orders: Low-high, Linear and Asymmetric
- No IPAT;
- No Half or Partial Fourier;
- No hybrid sequences;
- No rectangular voxels and no rectangular FOV;
- Only T2 and PD Weights are possible with SE sequences, due to long ET and the fact that all sampled echoes contribute the effective TE of the images.

Although not used often, radial acquisition can be quite useful in brain and joints imaging [5] to guarantee readable images when other method fail due to high amount of artefacts.

It is important to bear in mind that image contrast is slightly decreased by radial acquisition.[5, 17] Signal intensity fluctuation within same tissue and Streaking artefacts may occur, if a low number of blades is used (this will be particularly perceptible if the structures within the area of interest contain sharp edges (e.g. Bones and joints with irregular non-circular shapes)) (Figure 42).[8, 11]

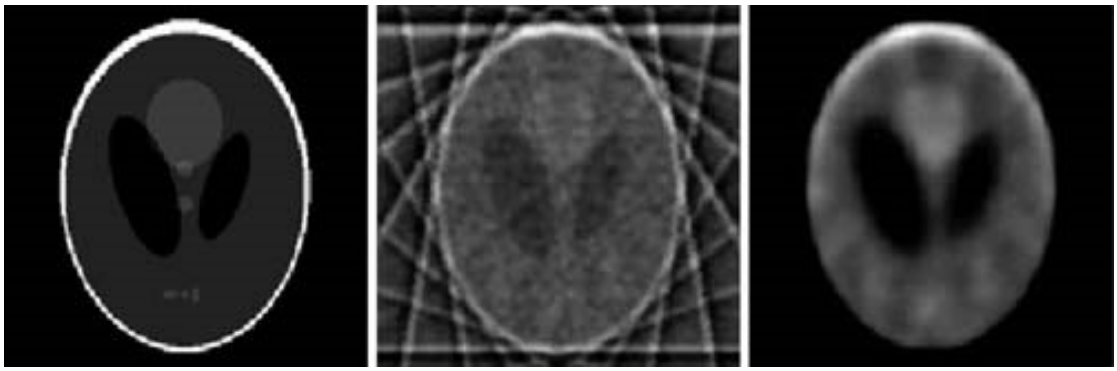


Figure 42: Right image: Virtual phantom; Middle image: Reconstructed phantom using radial acquisition with low number of blades. Severe Streaking artefact is seen throughout the FOV, particularly outside the area of interest; Left image: Reconstructed phantom with radial acquisition and adequate number of blades. Substantial contrast fluctuation within the same tissue and blurring of structures is seen within the area of interest.

11.1.6 Echo Planar Imaging (EPI)

Used for very fast acquisition sequences, like single shot sequences, EPI acquisition was in fact one of the first K-Space filling techniques suggested for MRI. Dating back to the primordial days of MRI, the concept of reading the MRI signal in an in-plane spiral direction came about following on the idea to use magnetic gradients as a way to spatially encode the protons and thus the signal itself.[3, 8, 11, 13]

The application of such concept took years to come into existence, due to technological limitations [8, 11], and it still needs improvement.

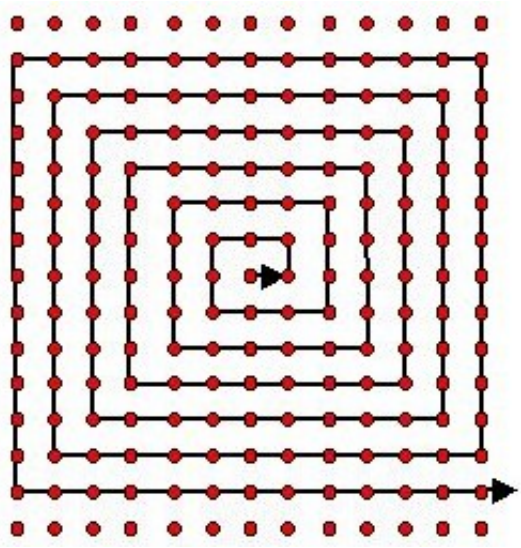


Figure 43: EPI acquisition. Adapted from [8]

Even though from an abstract point-of-view EPI looks simple, it's from a technical perspective very difficult to execute and full of limitations. This is why it isn't the mainstream acquisition technique, particularly in MSK imaging, but that may change in the future.

Unlike Cartesian or Radial acquisitions, EPI normally can only use one acquisition "line" per image (non-segmented EPI). This means that:

1. TR and ET need to be extremely long (long enough to sample a number of points that is at least equal to the number of pixels in the image);
2. Extremely fast and precise gradients are needed to enable filling the entire K-Space before the signal vanishes;
3. High BWs are fundamental in order to excite the protons and sample the signal quickly and correctly.

From a signal acquisition point-of-view there is always some "dead time", within every TR/TA. This is the time interval between the end of the ET, in a determined TR, and the moment when the slice/volume selection gradient is applied, in the next TR. In non-segmented EPI acquisition "dead time" is almost non-existent because the number of TRs is equal to the number of slices/volume times the number of NEX. This is not normally the case in other acquisition methods, where the time dedicated to signal acquisition is only a small fraction of the entire TA.

EPI acquisition is faster than other techniques mainly because it has a very long ET with a large number of echoes and so it requires much less TRs per image. Unfortunately, acquiring several echoes per TR has drawbacks:

- The high number of echoes in single ET leads to;
  - Extensive blurring;
  - High sensitivity to susceptibility and motion artefacts;

  - Low intrinsic SNR;
  - Limits the attainable classic/basic types of Weights to very strong T2.
- The high BWs produces;
  - Very low SNR;
  - Short ES that generates enhanced fat-tissue's signal.
- The ultrafast gradients produce:
  - Very loud acoustic Noise.

To compensate all these drawbacks:

- NEX/Slice Thickness needs to be significantly increased;
- In-plane resolution must be low.

On the other hand, because there are few excitation pulses and no refocusing pulses the amount of energy deposited in the scanned object is smaller, leading to a reduction of effective SAR.

This method of acquisition is therefore only used in very specific types of sequences that require very fast sampling and where resolution is not a priority (DWI, SWI, BOLD)

EPI is doable as single-shot or as multi-shot. Single-shot (non-segmented) EPI fills the K-Space in a single echo train. In multi-shot (segmented) EPI, K-Space is divided into to two or more regions and then one TR is used to fill each region. By using segmented EPI, the length of the ET is drastically reduced, mitigating the severity of susceptibility artefact and improving image quality.

The TA of multi-shot EPI is significantly longer than in single-shot EPI, but still fast compared with other acquisition methods.

The improved image quality, over single-shot EPI, and the still considerably fast TAs are the reasons why there is currently a significant amount of interest in multi-shot EPI in field of Research.

11.1.7 Volumetric acquisition

Volumetric Acquisition is often used when very high resolution of small structures is desired and when there is a need to generate 3D models or multi-planar reconstructions. It´s particularly useful for scanning structures which are small and complex or that have high contrast, in comparison with adjacent tissues (e.g. IAMS).[5, 11, 13, 28, 29, 32]

Voxels tend to be isotropic, to avoid an unattractive pixelated appearance (also known as staircase artefact/effect) in some reformatted planes.

As quickly mentioned before (see subchapter 6.5.2), volumetric (or 3D) scanning does not make use of "slices". Instead, the gradient normally used to truncate the excited volume, which corresponds to the slice, is used as a second Phase gradient during sampling.[5, 8, 11, 13, 28] This means that proton excitation needs to be contained in space in a different manner.

In volumetric scanning the excited volume is referred as a Slab. The dimensions of the Slab are the same as the FOV´s (when only one slab is used). This means two things:[7, 11, 13]

1. There is no physical gap between "slices" (adjacent voxels in the Z axis);

2. The lack of gap and the large volume of excited protons make Volumetric Acquisition much more vulnerable to susceptibility and interference-related artefacts (e.g. Cross-talk, fold-over, etc.).

The long TA of volumetric scans makes SE sequences not viable [8, 11, 13], with partial exception of 3D TSE with variable flip angle and SE-EPI (see subchapter: 8.2.3.4).[28]

Besides limitations in the type of sequences that can use Volumetric Acquisition, some loss of contrast and some blurring tend to occur as a result of the long ET employed.[7, 8, 11] ET must be long to keep the TA within an acceptable value.

In orthopaedic MRI, volumetric scans are less common than one would expect, being more often used in research than clinical exams. This can be explained by several factors, the most important ones being:

- Preference for high resolution SE sequences;
- Tradition in reporting by scrolling through same plane images Vs. Reporting using multi-planar reconstructions;
- Some significant loss in image contrast, sharpness and in-plane resolution;
- Higher prevalence of artefacts than in 2D scanning;
- Higher difficulty in manipulating parameters and adapting the sequence to the case at hand;
- Longer TA than 2D scanning in a direct comparison (not if the amount of clinical information obtained is considered or if multiple 2D scans with the same Weight are acquired within the same examination).

11.2 Parallel imaging

Parallel imaging (IPAT) techniques are used to greatly reduce the TA of a sequence without compromising image resolution. [3, 5, 7, 11, 22, 32, 37] This is done by intentionally creating fold-over artefact and then unfolding the artefact and using it to fill the information that is missing in K-Space.

It can also reduce SAR because it reduces the length or number of the echo trains used per image. [5]

IPAT requires the position of the correlated coil's channels to be in parallel with the Enc. Dir. (e.g. If two receiving coils, of 1 channel each, are positioned AP in relation to each other, then the phase encoding direction also needs to be AP for IPAT to work.)

11.2.1 SENSE

One of the first forms of IPAT developed was codenamed SENSitivity Encoding (commercial names: SENSE for Philips and mSense for Siemens. See page 139).[5, 7, 22, 32, 37]

SENSE achieves faster TA, by skipping N_{PE} in K-Space.[3, 5, 7, 11, 22, 32, 37] The number of N_{PE} is reduced by a factor defined by the operator. The maximum value that SENSE's Acceleration

Factor (AF) should be set to, is dependent of the number of channels present and their spatial relation.[5]

All coil's channels measure, simultaneously and separately, the entire region. The signal acquired by each channel is then compared with the others in order to correct for the reduction in N_{PE} and the signal misplacement (fold-over) that occurs.[3, 5, 7, 11, 22, 32, 37] Essentially, SENSE makes intentional use of fold-over artefact to obtain information of the whole FOV to avoid having to acquire the full amount of N_{PE}.

When the number of N_{PE} is reduced (and so the acquired FOV becomes "slimmer" than the original FOV) fold-over occurs (see subchapter: 14.2.1.1.1). This means that the signal from the area not sampled (N_{PE} not filled), of the original FOV, isn't lost but instead spatially misplaced on top of the acquired N_{PE}. By knowing, à priori, the sensitivity profiles of each coil's channel and their spatial relation to the FOV, it's mathematically possible for the scanner to correlate the data from all channels. This way is possible to separate the misplaced signal from the not misplaced signal. By doing this, SENSE can correct the signal misplacement and "unfold" the image into an image with an FOV equal to the original one and "without" any fold-over (if conditions are ideal). This unfolding process is done in Real Space, after the Inversed Fourier Transformation is applied.[3, 5, 7, 11, 22, 32, 37]

It is sort of like reading a transparent book. When closed the pages overlay on top of each other and it's difficult to read the words. SENSE knows how the pages overlay on one another and so it can open the book.

As an example: using two coils with one channel each and a SENSE AF of 2, only half the number of phase encoding lines are acquired to measure the same FOV, while maintaining resolution. This results in a reduction of the acquisition time by a factor of 2 and a reduction of SNR for a factor of $\sqrt{2}$. [5]

The higher the SENSE factor used, the less phase lines are acquired and more "layers" of fold-over occur. The more "layers" there are the harder it is to properly "unfold" the signal, resulting in artefact right in the centre of the FOV, if the process fails at any point. It is therefore important to keep the SENSE factor as low as possible or to increase the amount of co-relatable channels.

On Philips' scanners SENSE's AF is defined in decimal numbers (1.0; 1.1; 1.2…3.0…) which allows for "fine tuning" the degree of IPAT that the operator desires. This guarantees a good balance between TA and SNR when SENSE is employed.

Some AFs may generate blurring in the phase direction (increasing the pixel acquisition size) or alter the value of TR and ES. If the Enc. Dir. is not in-line with co-relatable channels a conflict will emerge and the sequence won't run. The sensitivity profile and spatial relation of the coils' channels is obtained through the acquisition of a Reference Scan[5] done prior to the acquisition of the sequences that use SENSE or CLEAR (see subchapter: 9.2.12).[5]

[5] The newest generation of Phillips' MRI scanner don't require the acquisition of a Reference Scan. Information regarding the coil's sensitivity profile is instead acquired during pre-acquisition phase of every sequence (in the same fashion as Siemens' MRI scanners).

On Siemens' scanners the factor of mSense follows a natural numbers order (1; 2; 3…). A Maximum Recommended AF is suggested based on the number of channels, covering the area of acquisition of the sequence that the scanner can correlate in the defined Enc. Dir. and the correlation method used (CP, Dual or Triple). The sensitivity profile and spatial relation between coils' channels is done in the preparation phase of the sequence.[22, 24, 37]

11.2.2 GRAPPA

Generalized Auto-calibrating Partial Parallel Acquisition (GRAPPA) is only available in Siemens' scanners and works in a similar fashion as SENSE, with the main difference being that the "unfolding" process in done in K-Space, instead of in Real Space. [7, 22, 37]

GRAPPA is more robust than SENSE, allowing for a higher Acceleration Factor (AF) (normally between 2 and 3) and less likelihood of severe fold-over to appear. [24] The drawback is that AF cannot be fine-tuned as SENSE. Only AFs using natural numbers can be employed.[24]

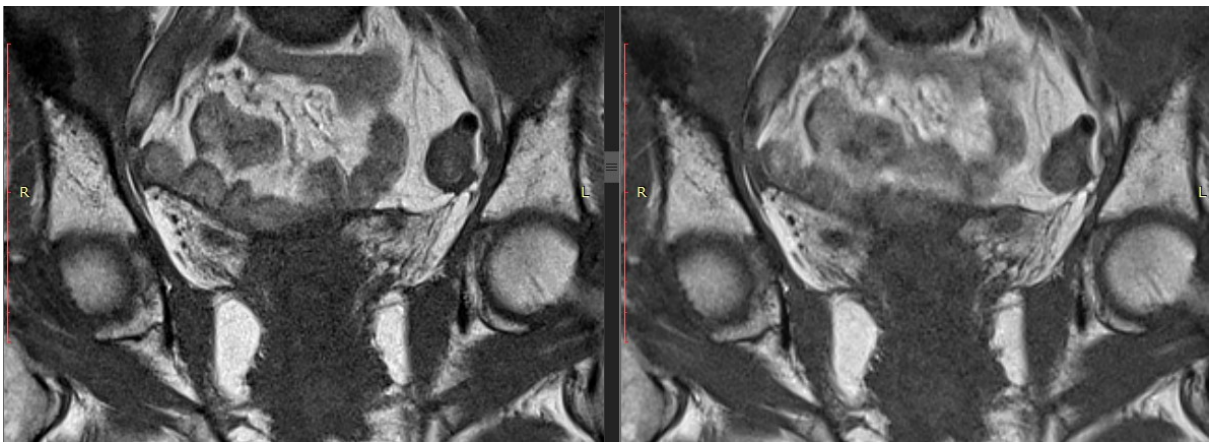


Figure 44: Increased noise from excessive AF denoted on left image. Left image AF=4; Right images AF=3

When artefacts occur, they tend to be significantly less severe than when SENSE is applied. If GRAPPA fails to work, two distinct artefacts may occur: curved lines (Figure 45) of artefact overlapping on the image and/or a large amount of noise (Figure 44) throughout the image.[37] This is often the result of applying an excessive Acceleration Factor for the case at hand. This is why Siemens's Syngo software has a Maximum Recommended AF parameter. Going beyond the Maximum Recommended AF is possible, but is not advisable in the majority of cases.

GRAPPA only works in an in-plane (one-dimensional) Cartesian acquisition, but there are currently other IPAT methods that work either in both Phase and Slice direction (bi-dimensional), by simultaneous multiple excitation of slices (MS-IPAT) or/and with radial acquisition. All these methods further decrease TA by allowing higher AFs, especially in 3D sequences, Dynamic scans, DTI and DWI. [37]

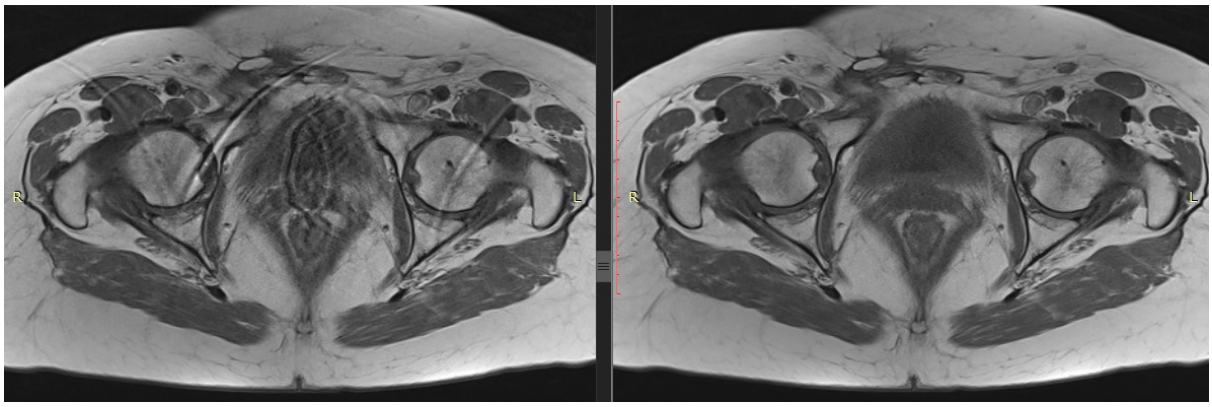


Figure 45: Same sequence acquired using different GRAPPA's Acceleration Factors. Left image showing GRAPPA related artefact (curved lines) from excessive Acceleration Factor for the number of coils. Left image's AF=3; Right image's AF=2; Enc. Dir.: AP; Body and Spine coils used. SNR and TA equalized through reduction of oversampling.

12 Breath-Hold scans

Most of MSK MRI examinations consist in scanning body parts where the motion caused by the patient's breath won't usually create severe amounts of breathing or cardiac motion artefact.

Breathing artefacts are quite prevalent in thoracic and abdominal scans. In MSK examinations they can still be routinely seen on: pelvis/hip, ribs, shoulder, clavicle, sternum-clavicular joints, neck, whole spine, and brachial plexus exams.

In most cases, like in spine or pelvic examinations, the amount of artefacts caused by breathing is significant but easily controllable by choosing a proper Enc. Dir. and making good use of saturation bands. This ensures, respectively, that the artefact is constricted to a small area of no interest or it's minimized through saturation of the area where the artefact originates from.

Clavicle, ribs and sternum-clavicular joints are more problematic structures to scan because it's not a case of having a movable structure in the vicinity of the area of interest; it's a case that our area of interest is actually moving along with the patient's breath. Breath-Hold acquisition may be a quite useful tool in such instances.

Although the acquisition of information while the patient holds their breath is an almost unused technique within MSK MRI, there are still some situations (e.g. scanning ribs, chest wall, abdominal wall, etc.) where it may provide significant improvement to image quality (through artefact reduction), in relation to regular Cartesian acquisition and even Radial acquisition.

Breath-Hold scans consist in single of multiple quick acquisitions (in the order of 15s to 25s and regulated by the total TA divided by the number of Concatenations) while the patient holds their breath.[11, 13] The sequences used (e.g. HASTE; Flash; GRASE; TSE with long TF, etc.), have low or medium resolution and employ long ET to keep TA very short.

Alternatively, acquisition regulated by a navigator (placed midway between the right lung and the liver) or another triggering method are also available in most scanners.[5, 11, 13] These methods require regular breathing throughout the entire scan for satisfactory results.

13 Filtering

A filter is a mathematical algorithm to which all the data (pre, during or post image reconstruction) is passed through, in order to amplify and/or undermine certain characteristics of the data. A good image filter is always subtle and corrects/modifies the data without compromising/distorting the information that the image provides to the observer.

The list of filters can are used in MRI is very extensive, an each one has a specific purpose, but all of them either correct a flaw in the reconstruction process or manipulate the information contained within the image. Some filters cannot be used together or without some specific parameter being active, but is unusual.

It is important, for the operator, to know the most commonly used ones to best take advantage of their usefulness.

13.1 Smoothing filter

It's a low band-pass filter that enhances the contribution of the low frequencies to the image, during the reconstruction process. It produces images with subtle enhanced contrast in detriment to detail. This occurs because the information related to the image's contrast is retained in the low frequencies that compose the sampled signal.

It can also be used to partially hide noise. When used in excess it will blur areas of contrast transition.

13.2 Sharp filter

It's a band-pass filter for high frequencies, instead of low frequencies. It produces results opposite to a low band-pass filter. Small details are enhanced, including noise. It can result in deterioration of contrast through significant increase in noise, disturbing the image's CNR.

It can be used to sharpen areas of contrast transition. When in excess it will produce a significant amount of noise and intensity heterogeneity within the same tissue type.

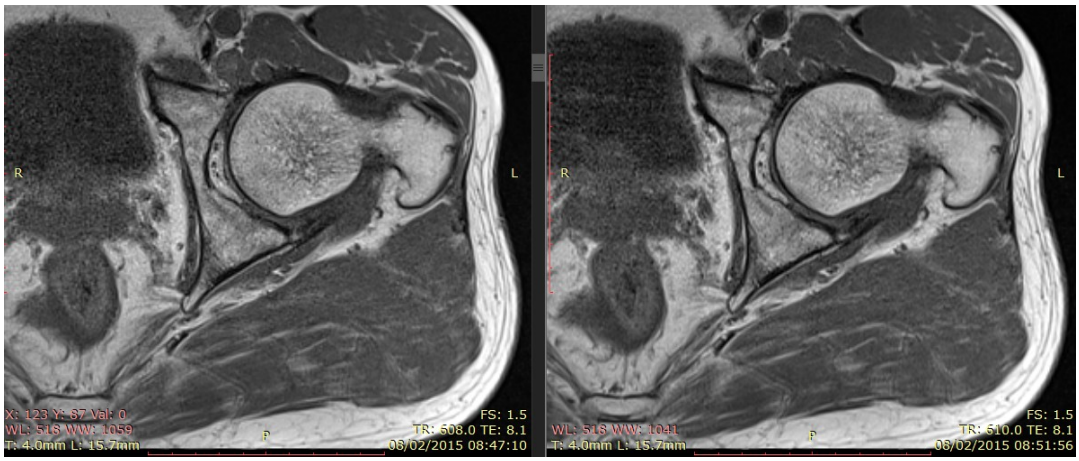


Figure 46: Left image not filtered Vs. Right image filtered with a smoothing filter (mild intensity). The effects of a smoothing filter are more easily perceived within large areas of supposedly equal intensity (e.g. Bladder, muscle, bone marrow, etc.). A sharp filter will be more apparent in areas of tissue transition and enhanced long thin structures (e.g. Cortical bone, ligaments, vessels, etc.).

13.3 Geometry Correction

It corrects geometrical distortions that occur at the edges of the FOV (pillow-shape distortion). The distortion is produced by having a square-shaped image/FOV in the non-square geometry of B_0. Correction is performed by reshaping the image into a circle within the square that is the FOV in order to counteract the distortions that are produced..[5, 24]

13.4 Raw filter

This filter supresses oscillations near the edges. These oscillations get erroneously registered within the outer lines (where information regarding motion is kept) of the Raw Data that is encoded within K-Space. The correction is done by weighing specific lines of the Raw Data's outer lines.[24]

13.5 B_1 filter

Reduces the signal oscillation produced by dielectric resonances that are generated at field strength equal or higher than 3T. It should be always used if 3T scanners, but is more important for large FOVs.[24]

13.6 Elliptical filter

Similar to Elliptical Sampling Acquisition (see subchapter 0) but it doesn't reduce TA by avoiding acquiring K-Space's corners. Instead and prior to image reconstruction, a shutter is applied to the corners of K-Space setting the value of the Raw Data information, contained in the corners, to zero. SNR is visually improved in the order of 10% without loss of Spatial Resolution in the reconstructed images.[24]

13.7 Ringing Filtering

Filter applied during image reconstruction to reduce the severity of oscillation in signal intensity in areas of sharp contrast transition. It generates a bit of blurring in the images, reducing spatial resolution.[5]

It is mostly used in sequences that use EPI acquisition and/or volumetric acquisition.

It can be applied in rectangular form, affecting the entire K-Space evenly, or in an elliptical fashion, where the edges of the K-Space are filtered more than the centre. This last form of this filter generates significantly more blurring, visually "improves SNR", and gives a smother look to tissue's textures.[5]

14 Technical challenges

14.1 Signal-to-Noise Relation and Relative Signal-to-Noise Relation

Technically speaking, SNR is nothing more than a mathematical relation between the amounts of true signal and noise sampled, but it's an important factor in any device that converts analogic signal into digital information. In MRI, SNR is one the cornerstones of IQ and sequence optimization. Finding the lowest amount of SNR necessary to get adequate IQ, within a specific context (patient's morphology; coils used; field strength; spatial resolution; sequence; image's Weight; etc.), is a gruelling, but essential, try-and-error based task, for optimizing any given sequence.

While it is "fairly easy" to calculate the SNR of a given image, it is impossible to reliably calculate the SNR of any sequence. There are too many factors that influence the intensity of the sampled signal and to make things worse, many of them have unknown values that change from case to case (e.g. Patient's weight). For this reason, the actual SNR of a sequence is never provided by the MRI scanner. Instead, another parameter is provided: Relative SNR (rSNR).

It is normally provided in the form of a percentage or a decimal non-dimensional value and, by itself, it says nothing about the actual SNR of the sequence. What rSNR provides is a mathematical ratio between the assumed value of the original sequence (which is always defined as 100% (or 1)) and the value of that sequence plus the alterations introduced by the operator, through manipulation of parameters.

If after manipulation of parameters the sequence is saved, the rSNR is automatically reset to 100% or 1 again (regardless of what value it was before). For this reason rSNR cannot and should not be assumed as a reliable indicator for SNR (this mistake is one of the most common errors that operators' make and it either leads to grainy images or to excessively long scanning times). Nevertheless, rSNR is a very useful point of reference to assess if the parameters' changes introduced, by the operator, will provide lower, equal or higher SNR than the original sequence. Additionally, if a given sequence is optimized (SNR wise) for a specific context (defined patient morphology (e.g. bariatric patient); type, position and number of coils/channels; sequence used; image's Weight; etc.), rSNR can be used to adjust the optimized sequence to a new context (e.g. decrease FOV to adjust it better to a smaller body part and then adjust rSNR back to 100% (or 1) to guarantee adequate SNR).

The parameters that affect rSNR change depending on the manufacturer. This doesn't mean that if a parameter doesn't change rSNR it won't change SNR. It just means that manufacturers use different equations to calculate rSNR and some of them ignore the influence of certain parameters to SNR (e.g. Siemens' rSNR equation ignores changes in TR, TE and FAng even though these parameters affect the intrinsic SNR of any sequence).

14.2 Image Artefacts

Artefacts are simply put, anything that appears in an image that provides false information about the scanned object. This is a very simplistic way to define image artefact, but because the list of MRI artefacts in so extensive and diverse in both origin and appearance, in practice, it's probably the best way to define image artefact.

MRI is unquestionably the medical imaging modality where artefacts are more prevalent. It's one of the drawbacks of MRI's flexibility/potential. Considering the massive amount of changeable factors at play, during acquisition, it's only normal that errors and conflicts occur, generating unwanted artefacts.

Varying in nature, severity and appearance it's hard to keep track on all artefacts that may present on a MR image. Things get even worse because, depending the type of image desired, what causes an artefact in a given image, may be exactly what we need to generate a different type of image. For example, chemical shift can be a subtle but important artefact in a T1 image of a joint, but without it fat saturation techniques wouldn't be feasible.

Virtually it can be said that every MRI image has artefacts, what matters is their severity and if it affects the information that will be extracted from the image.

14.2.1 Acquisition-related artefacts

14.2.1.1 Aliasing artefact

Artefacts that result from the non-compliance with the Nyquist Theorem (see subchapter 6.5.5), including improper truncation of K-Space.

14.2.1.1.1 Fold-Over artefact

Also known as Wrap-around artefact, Fold-over is the result of improper or insufficient truncation/saturation of the signal outside the FOV. It occurs always and only in the Phase Enc. Dir. in 2D scans and in the Phase and/or Slice Enc. Dir. on 3D scans.[5, 7, 11, 13]

It appears in images as overlapping, of structures located outside of the FOV, into the FOV. Structures located on one side of the FOV will appear overlapping on the opposite side of the image. [5, 7, 11, 13]

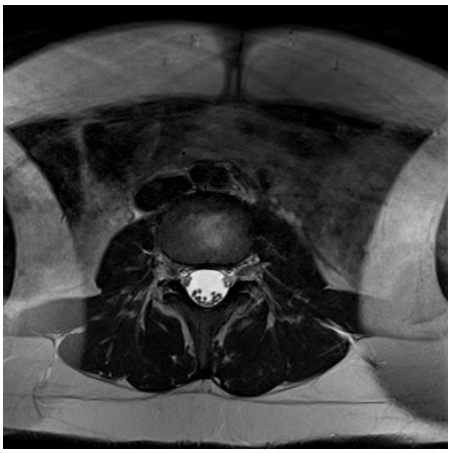


Figure 47: Axial T2 of lumbar spine. Fold-Over artefact.

The expression: "trying to square a circle" exposes the physical "impossibility" that creates this artefact. The acquired signal is electromagnetic radiation, so in essence a complex wave. Waves have a one dimensional nature and extend themselves periodically in space until infinity. When K-Space is filled, waves are constricted within a finite space. This is done by truncating the wave using a secondary

modulating wave that interacts constructively inside the FOV and destructively outside of the FOV. This secondary modulating wave is a known complex wave formulated exclusively to truncate the signal, if done incorrectly Fold-over artefact occurs. Truncation can be improved with through oversampling methods and saturation of the protons outside the FOV.

14.2.1.1.2 Ringing/Truncation Artefact

Also known as Gibbs´s artefact, Ringing is a form of Aliasing that presents itself as subtle alternating bright and dark thin lines, adjacent to boundaries of high contrast. These lines possess the same shape as the boundary line and can appear in either frequency or phase direction. It appears more often in the Phase direction due to the tendency for this encoding direction to be the smaller one in the matrix used. It is most commonly seen in sagittal views of the upper spine, on water-phantom images and on images of the bladder.[5, 7, 11, 13]

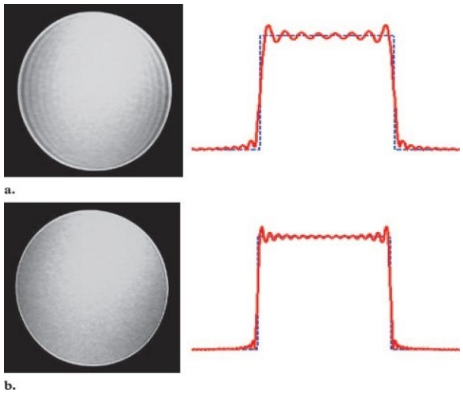


Figure 48: Ringing artefact seen on top image as bright and dark lines at the edge of the water phantom.

It´s caused by insufficient sampling of all the frequencies that compose the signal´s wave, resulting in improper truncation in K-Space.[5, 7, 11, 13]

It can be corrected by improving the acquisition matrix in the encoding direction where the artefact is seen.[5, 7, 11, 13]

14.2.1.2 Chemical Shift Artefact

The hydrogen protons in fat tissue have a Larmor Frequency slightly lower than protons in water. The difference in value between these two frequencies is called chemical shift. The difference tends to be very small and, as a result, displacement of both signals can often occur during sampling. [5, 7, 11, 13, 24, 30, 31]

When both signals are displaced, Chemical Shift artefact occurs (Figure 49 & Figure 50). It presents as two lines of hypo and hyper signal (opposed to each other in the frequency direction), in areas of boundary between water and fat tissue (Figure 49). This artefact depends solely on the numerical “distance” between the resonance frequencies peaks of these two tissues and can be manipulated by variation of B_0 (increase in the field strength increases artefact substantially) or by varying the rBW.[5, 7, 11, 13, 15, 24, 30, 31]

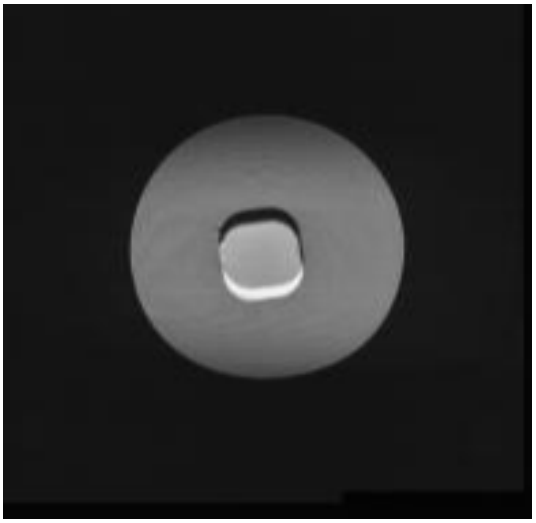


Figure 49: Chemical Shift artefact demonstrated in a MRI phantom. False lines of hypo and hyper signal in bounderies between water and fat. Extracted from [5]

Higher rBW will reduce Chemical Shift artefact because the range of frequencies sampled per

pixel increases allowing better discrimination of small variations in frequency values and so less possibility of miss registering fat for water, or vice-versa. [5, 7, 11, 13, 15, 24, 30, 31]

The importance of Chemical Shift artefact relates to the fact that both water and fat are very common type of tissues, inside the human body, but that their presence in a specific place has a very different meaning. Moreover when these tissues are adjacent to each other a Chemical Shift artefact can compromise proper delineation of the boundary between tissues/structures and promote degradation of image's sharpness.

On sequences that use selective spectral saturation techniques of either Fat or Water, Chemical Shift artefact is less significant because one of the tissues is saturated [7, 11, 15, 38], yet it still occurs (Figure 66). This is more prevalent in large FOVs and volumetric scans.

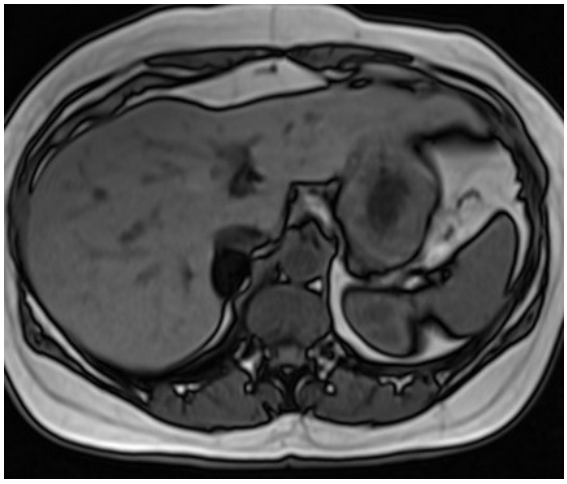


Figure 50: Axial T1 localizer of abdomen. Out-of-Phase image with characteristic secondary nature Chemical Shift artefact in water-fat boundaries.

On MSK examinations the Chemical Shift per pixel should be set between 1.0 and 2.5 pixels [38] to prevent occurrence of severe artefact. If metal implants are present then is should be around 0.6 pixels [7] (see subchapter 21.1.2) to minimize metal-related artefact (see subchapters 0 and 21.1).

14.2.1.2.1 Chemical Shift artefact of secondary nature

It's a form of Chemical Shift that results from sampling both water tissue and fat tissue when they are in opposing Phases. It's appears as a distinctive dark line all around water/fat boundaries.[7, 11, 13, 30] It is a characteristic of Out-of-Phase images (see subchapter 10.4) but it can also be seen in other images if certain TEs are selected (especially in GE sequences because of the lack of Phase refocusing during echo creation).

It can be corrected (if it's and undesired side effect) by changing TE to an In-Phase TE (Table 8). Out-of-Phase and In-Phase TE's values change with field strength.[5]

14.2.1.3 Moiré Artefact

Moiré is also known as Zebra Stripes artefact because of its appearance as dark stripes overlapping the edges of the FOV. It's caused by significant RF inhomogeneity and so is often more easily seen in large FOVs and where scanning structures that are too close the scanner's transmitting coil/bore's wall.[11]

The only solution is to move (if possible) the scanned structure away from the area where the artefact is seen.

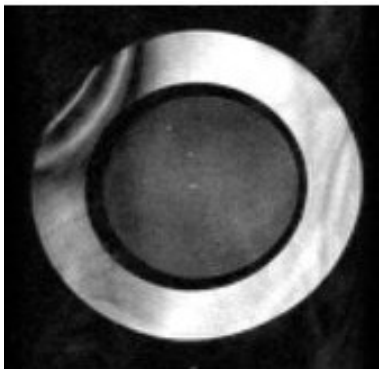


Figure 51: Moiré Artefact at the edge of FOV.

14.2.1.4 Magic Angle artefact

Magic Angle artefact refers to a curious signal enhancement observed in specific tissues/areas, when the effective TE is low (less than 35ms). It seen in at types of Weights within exception of T2, in both SE and GE sequences.[39]

The effect is generated in tissues that have tightly compacted water molecules within collagen fibres (e.g. Tendons and Ligaments) and are angulated around 54.74º (hence its name) in relation to B_0.[39] Under these conditions the water-bound hydrogen possesses a short T2 and T1 and produces the hyper signal observed.

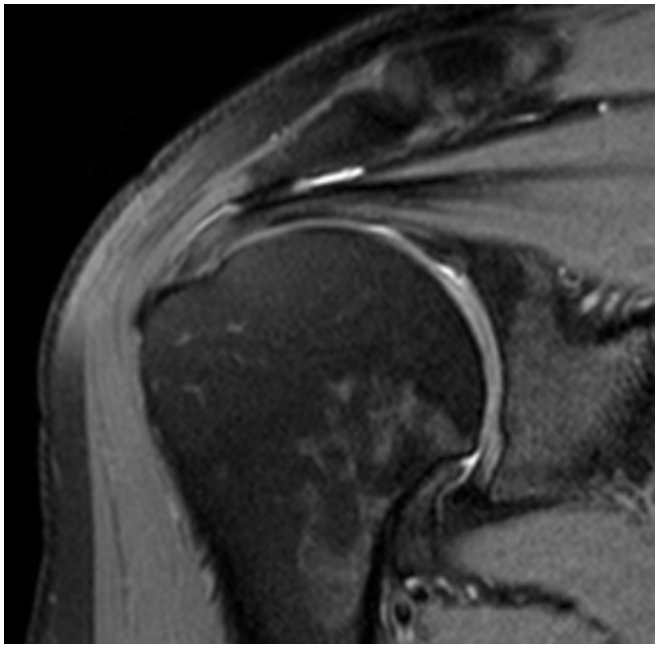


Figure 52: Coronal PD_FS. Magic Angle artefact (or effect) seen in the insertion of the supraspinatus muscle into the humeral head. TE=33ms

It's seen often in around joint, with particularly incidence in the shoulder, knees, ankle and hip.

Correction is only achieved by changing the angulation of the object in relation to B_0, but that is not possible in many situations. Using longer TEs than usual (particularly in PD weighted images) can mitigate the artefact to some extent.

14.2.1.5 IPAT related artefacts

The artefacts that can occur from using IPAT depend of the method used, the AF used, and potential patient/coil motion during acquisition. This topic was already covered in subchapter 11.2.

Correction of these artefacts can only be done by reducing the AF or the use of IPAT altogether.

14.2.1.6 Cross-talk artefact

As mentioned before, Cross Talk between slices may occur if there is no physical gap between adjacent slices, if Sequential Slice Acquisition is used. It will definitely occur if the slices are not parallel and instead cross each other at some point in space.

This artefact only occurs in multi-slice acquisition and it's very common in axial slices of the spine that employ multiple stacks of slices and in multiplane localizers.

It results from destructive interference during slice selection/excitation.[5, 7, 11, 13] The non-parallel alignment of the slice ensures that the slices are acquired out-of-phase in relation to each other and so the interference is always destructive.

A single or multiple parallel lines of dark (void signal) appear in the direction(s) of the crossing of the slices. The central line is always the thickest.

This artefact can only be corrected by changing the angulation of the slices/stacks in order to move the point of crossing to an area outside the FOV.

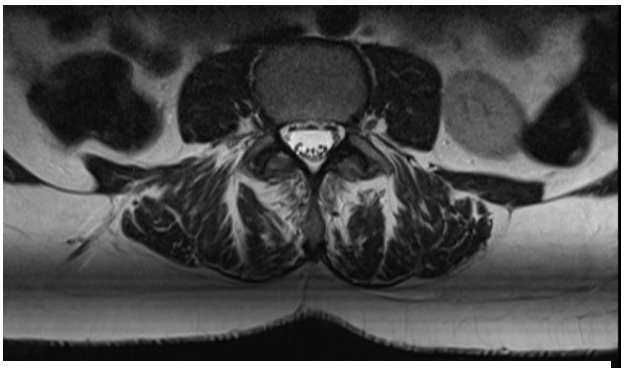


Figure 53: Axial T2 of Lumbar spine. Cross Talk artefact in an axial slice of the Lumbar spine

14.2.1.7 Dielectric Shading artefact

It present as one or multiple areas of generalised darkness, separated by an area of normal brightness, within the FOV. It's more commonly seen in field strengths equal or higher than 3T and in large FOVs.[5, 11]

It is caused by the amplitude variation observed in the transmission RF pulse, in different areas of the FOV, and produces oscillations in the degree/intensity of excitation that the protons of those different areas experience.

It can be corrected by using simultaneous parallel out-of-phase RF multiple transmissions.[5] Most current technology already uses multi-transmission, but older scanners and coils can still be solely single transmission. Dielectric Shading artefact (also known as simply: Shading) can also be mitigated by using brightness Normalization (see subchapter 9.2.12).

Sometimes Shading artefact may be produced by running the sequence while some of the channels or coil (in multi-array coils) are either turned off or faulty. Is this situation, localized graininess may also be seen in the darkened regions of the FOV. Other times the signal picked up by nearby coils/channels may correct the graininess. Regardless of the case, is fundamental that the operator check if all the desired coils are turned off and then rerun the sequence. If the problem remains even though the MRI system indicates that the coil is turned on, it will likely be that the coil/channel is faulty and need of repair.

14.2.2 Patient-related artefacts

14.2.2.1 Motion artefacts

It consist in a spatial miss-registration of signal in the Phase Enc. Dir. caused by sampling multiple times the same area of an object in different lines of the matrix.

The artefact appears as generalized blurring (if the degree of motion is small and/or quick) or repeated lines of the same body boundary (if degree of motion is large and/or slow). The severity depends on:

- The sequence's sensitivity to motion;
- The angular relation between the direction on the motion and the Phase Enc. Dir.;
- The degree of motion.

Cartesian acquisition is much more sensitive to Motion artefacts than other acquisition methods due to regular and linear acquisition of parallel lines of Phase and due to long TA.

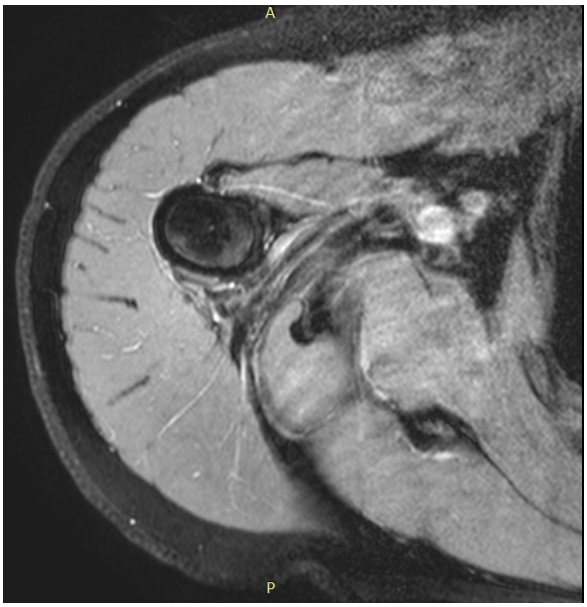


Figure 54: Axial PD_FS of shoulder. Motion artefact is seen.

Under-sampling of the edges of the K-Space can reduce Motion artefacts (this is actually one characteristic of Radial Acquisition (see subchapter 11.1.5) and of Motion Smoothing (see subchapter 9.2.9)).

The best things the operator can do to prevent Motion artefact are:

- Proper fixation of the body structure scanned;
- Request the patient's cooperation during the scan;
- Use sequences with short a TA;
- Use radial acquisition.

14.2.2.1.1 Breathing artefacts

Breathing motion is a specific form of Motion artefact (regular Motion artefact) and appears as streak lines in the Phase Enc. Dir. Its severity is dependent of the sequence used (normally the slower the sequence the worse the artefact), the rate and deepness of the breath of the patient. It can be solved in some cases by making the patient perform multiple breath holds during acquisition (on fast sequences) or in the large majority of MSK examinations by choosing a Phase Enc. Dir. that is orthogonal to the artefact's direction. This last option won't reduce the artefact but instead constrict it in smaller area of the image and hopefully away from the area of interest.

Alternatively, Radial Acquisition can be used.

Placing a Sat Band in the area where the motion is generated will minimize it.

It's important to stress, that in the majority of the cases, multiple motion reduction techniques must be used together.

14.2.2.1.2 Non-regular motion artefacts

Difficult to predict and hard to prevent against, non-regular motion appears as generalised blurring in the entire image or as streak lines near areas of high contrast between two distinct tissues (e.g. Cortical bones and soft tissue/bone marrow). It can be limited to just a few slices, or extend to all slices, depending on the characteristics of the acquired sequence and the severity of the motion.

Normally it's caused by the patient moving the entire structure, so there is no point of origin for the motion.

The best course of action is to repeat the sequence after alerting the patient for the need to keep still, or instead change the sequence to a radial acquisition mode (if possible).

Reducing the length of the sequence is advisable (even at the cost of spatial resolution or SNR), especially if the motion is involuntary and/or the sequence is lengthy.

14.2.2.1.3 Flow artefact

In-plane flow artefact is the result of building-up Phase errors in protons moving along a gradient. The build-up of errors occurs because the gradients are set for stationary protons at a specific TE and do not accommodate for Phase variations that occur by moving spins along an electromagnetic gradient. Applying in-plane flow compensation, making good use of Sat Bands and/or even using radial acquisition are the best solutions for the problem. A large number of measurements or the use of Asymmetric Profile Order can also help in mitigating the artefact.

This artefact can also occur in through-plane direction (slice direction) as a result of influx of excited protons into the FOV. In cases of this nature, correction is best achieved with a combination of slice direction flow compensation and Sat Bands positioned adjacent to the FOV (in the region of influx).

Artefact severity increases with the increase of the field strength. This is particularly true for flow artefacts. Transmit/Receive coils also contribute for this type of artefact. [5]

The severity of this artefact is highly dependent on the intensity of the flow and the direction of the flow in relation to the readout direction. The higher the intensity of the flow the more likely will artefacts appear. As an example, CSF flow around the cervical region is much more intense that in the thoracic or lumbar region, leading to the appearance of

14.2.2.1.4 Ghosting artefact

Ghosting is a form of Flow artefact that normally occurs due to vessel pulsation. It appears when a vessel, with a strong blood flood, crosses an image in an orthogonal orientation. Multiple Ghosting artefacts may appear in the same image, but each one will have a single point of origin that the respective line will cross at some point in its path.

It's particularly common in axial images, due to the orientation of the bodies' major vessels, Ghosting artefact present itself as a straight stripe of multiple pseudo-vessels that always runs parallel to the Phase Enc. Dir. and extends itself throughout the entire FOV. The pseudo-vessels are fake copies (ghosts) of the real vessel and are created by miss registering, in the Phase direction, the location of the real vessel, each time it pulses.

This artefact is more evident in high contrast images due to the signal intensity difference between the vessels and surrounding tissue.

The severity and appearance of the artefact can change slightly depending on the strength and frequency of the pulsation:

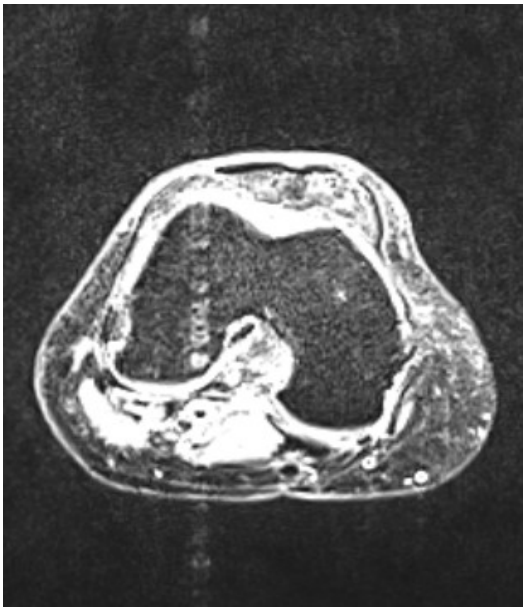


Figure 55: Axial PD_FS of the knee. Ghosting artefact is seen originating from the popliteal artery. Anterior-Posterior Enc. Dir.

- A strong pulse will create a strong artefact, leading to very defined circular shapes;
- A weak pulse will create a subtle artefact with poorly define circles;
- Quick pulsation will create ghosts with small intervals between each other. If the frequency is high enough, the ghosts will overlap and generate a blurred line;
- Slow pulsation will create ghosts with big intervals between each other. If some situations only a few ghosts will be visible;
- If through-plane Flow artefact is also present (which is likely), the ghosts of the same line can present themselves with variable signal intensity.

Ghosting can be partially corrected by turning on Flow Compensation (see subchapter 9.2.10) in the both in-plane and through-plane direction.

Changing the Enc. Dir. will only change the direction of the artefact, which in some cases will be enough to ensure that the artefact doesn't overlap onto the area of interest (e.g. axial images of the knee).

Alternatively, improving SNR, choosing Asymmetric Profile Order, or changing the averaging mode to: long, can significantly obscure the artefact.

Ghosting is one of the few types of artefact that tends to get more prevalent and severe with optimized sequences. This happens because optimized sequences rely on using only the smallest amount of SNR (and K-Space lines) necessary for adequate IQ, in order to keep TA low. This strategy often does not provide enough measurements for Flow Compensation, long averaging mode and Asymmetric Profile Order to be effective in obscuring Ghosting artefact.

14.2.2.1.4.1 Non-physiologic Ghosting

Although it's most often caused by vessel pulsation, Ghosting can also be caused by K-Space modulation. In this case, the artefact is not related to motion but with K-Space geometry. It will present itself as one or two ghosts of the entire scanned structure, next or partly overlapping on structure. In such cases, it becomes an equipment-related problem and quality control tests must be performed by a qualified MRI engineer.

14.3 Susceptibility artefacts

Magnetic susceptibility relates to how much a material, when placed within a magnetic field, interacts with the field. This means both a change in the material's magnetic properties and a disturbance of the field's strength. Simply put: how likely is for a material to become a magnet when placed within the vicinity of another magnet? If the answer is something other than "practically none", then we might be faced with a very problematic type of artefact.

14.3.1 Susceptibility artefact

When B_0 is not homogeneous throughout the entire FOV, some protons won't find themselves under the resonance conditions and won't get excited as much (or not at all) by the RF pulses. This results in small to medium darkened regions, because the signal from the protons is not as strong as it should be.

Susceptibility often occurs (when the FOV is large) at the edges of the FOV (but not solely at the edges) because B_0 is more inhomogeneous at the ends of the bore and in areas close to the bore's wall, than at the centre of the scanner/B_0.

This artefact can be partially corrected by applying some pre-acquisition and pre-reconstruction filters. Nevertheless, Susceptibility artefact at the edges of the FOV will be seen when the size of the FOV becomes significantly large. The exact size of the FOV, when this artefact starts to appear, is dependent on the scanner itself and on the sensitivity (to B_0 inhomogeneity) of the sequence used. An FOV larger than the mentioned size will add very little diagnostic information since, the signal will either be null or partially corrupted.

Tests should be carried out by the operator to find out the maximum FOV's size, for each sequence, when susceptibility artefact has yet to present itself. The results should be used as a limit for the FOV size when optimized the sequences.

1.1.1 Shim-related artefacts

Shim is used to correct B_0 inhomogeneities. It can be either passive (through the use of Shim Plates) or active (Shimming Gradients). When shimming is not performed or performed incorrectly Susceptibility artefacts (see subchapter 14.3.1) and/or improper tissue saturation/enhancement occur.

Improper saturation of fat tissue is the most common type of shimming-related artefact and probably the most common image artefact seen in MSK examinations.

Active shimming can be performed automatically, by the scanner, or manually, by the operator.

Spectral tissue saturation methods are very dependent on Shim to work adequately and in such cases manually setting the Shim's volume is advisable.

The Shim volume should avoid covering too many areas of air-tissue transition to work correctly.

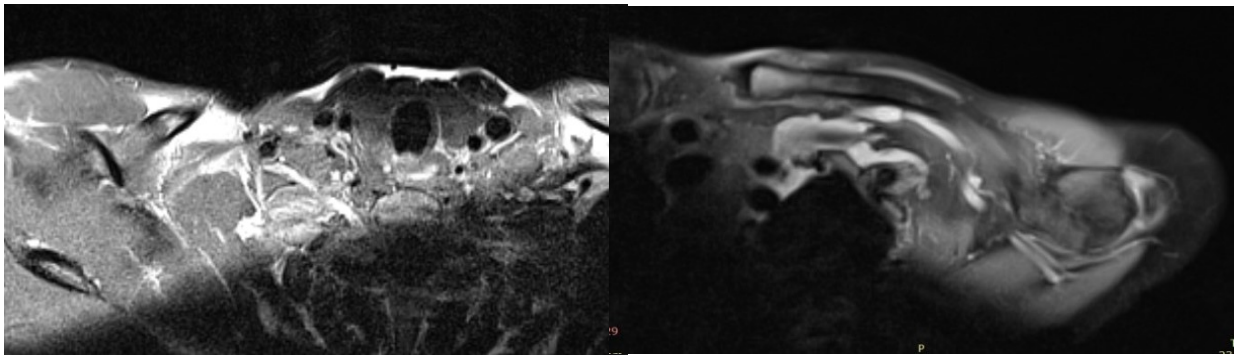


Figure 56: Left image: Axial PD_FS of right clavicle. Automatic shimming was used, leading to incorrect fat saturation. Right image: Axial PD_FS of left clavicle. Manual active shimming was used. Improper fat saturation artefact is almost non-existent.

14.3.2 Metal-related susceptibility artefacts

The presence of non-biological implants is quite common in today's clinical setting and unfortunately the large majority of non-biological implants are either metal-based prosthetics or electronic devices. From an image quality point-of-view none of this is good, because the large majority of these implants possess magnetic properties strong enough to disturb/distort B_0. Consequently the Larmor Frequency of protons, located in the vicinity of the implant, also changes, which inevitably leads to artefacts in the images.[5, 7, 11]

These artefacts appear as darkened areas, of signal void, sometimes also associated with line(s) hyper signal surrounding it. Their severity can range from almost insignificant to catastrophic corruption of the image's information, depending on a large list of factors that may be at play during signal acquisition. The best and only reliable way to avoid these artefacts is to remove the object that causes the B_0 disturbance from the vicinity of the FOV. If this is not possible, and unfortunately this is the case in most situations, then it's important to consider the use of Metal Artefact Reduction Sequences (MARS) (Figure 58).

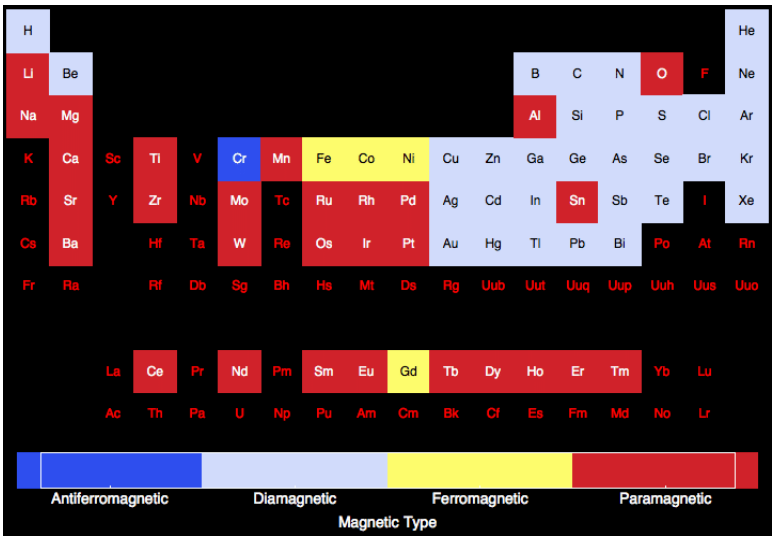


Figure 57: Magnetic properties of elements in the Periodic Table of Elements. Extracted from [6]

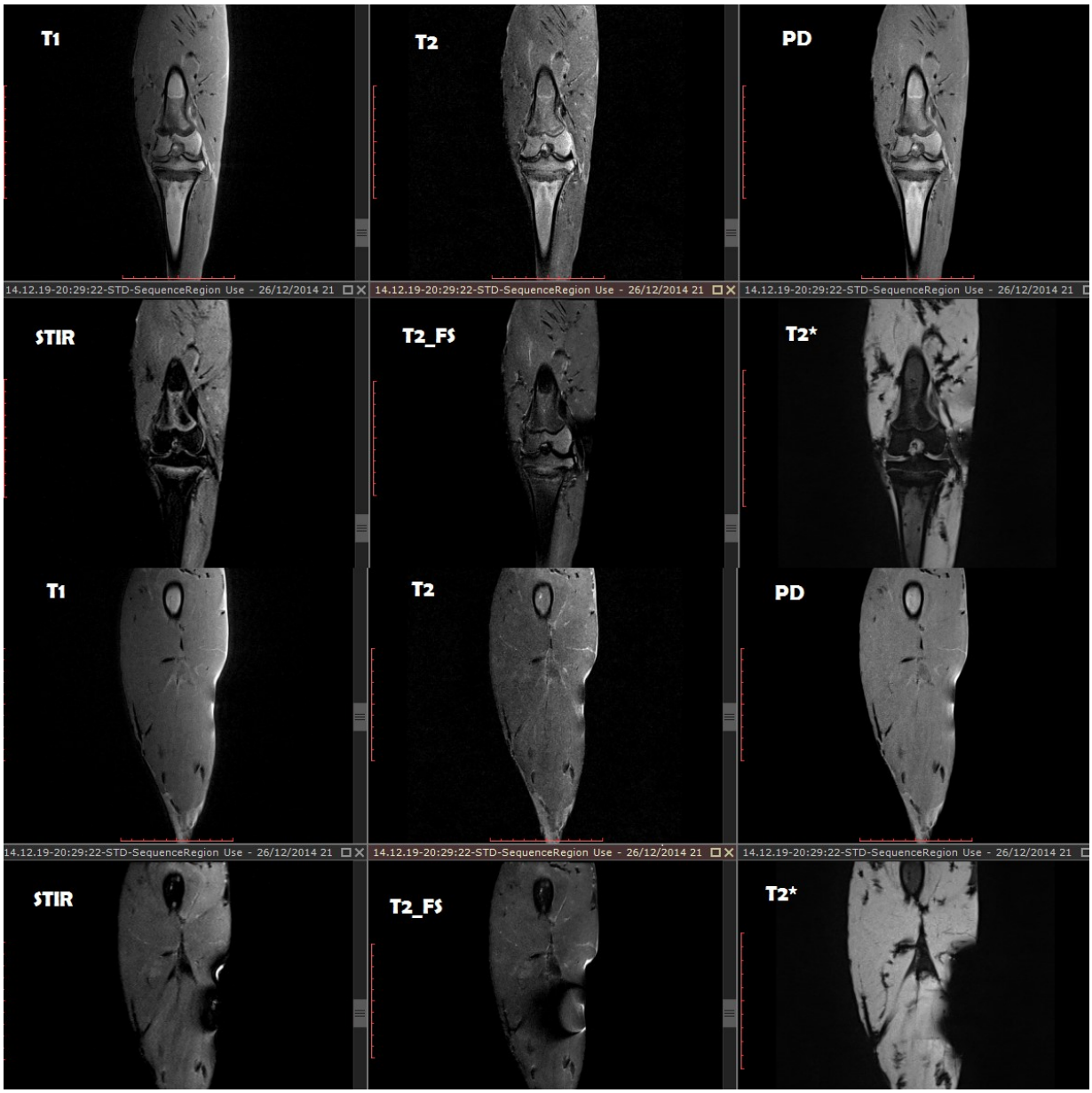


Figure 58: Severity of Metal-based artefact for different MRI Weights using MARS. Images labelled for Weight. Even with parameter manipulation to reduce metal artefact the differences between Weights are obvious. Important to notice the improper fat saturation in the knee joint on T2_FS[6]. This is not seen in any other Weight, even though the area of void signal is bigger on STIR and T2*. Metal Artefact produced by placing an 8 centimetre length 0.5mm diameter iron rod in contact with the phantom's skin. TSE sequences used for all Weights apart from T2* (GE).

[6] T2_FS -T2 sequence with Fat Saturation

There are many factors that influence how severe the metal-related susceptibility artefacts will be, but the most important ones are: [5, 7]

- Size of the implant – Affects the degree of B_0 disturbance. The greater the size, the worse the artefact is;
- Composition of implant (Figure 57) – Affects the degree of B_0 disturbance/susceptibility. The more ferromagnetic (Iron, Nickel or Cobalt) or paramagnetic the materials are, the worse the artefact is;
- Position of the implant within B_0– Positioning the implant in locations where it experiences significant Torque (e.g. close to bore's wall) will increase the artefact's severity (It's a factor that normally cannot be changed).
- Angular relation and between B_0's vector and the long axis of implant – Positioning the implant in an orientation) that promotes the occurrence of Torque (normally diagonal to B_0's vector will produce greater artefacts (It's a factor that normally cannot be changed).
- B_0 Strength – It affects the proton's Larmor frequency and the intrinsic SNR. Higher field strengths produce worse artefact;
- Type of sequence used – GE sequences are much more sensitive to the artefact than SE. TSE is better than SE and TIR. Spectral saturation is much more sensitive than IR. EPI, volumetric and Radial acquisition are more sensitive than Cartesian acquisition;
- Receiving BW – Affects the degree of signal/frequency shift. The higher rBW is the smaller the artefact;

14.4 Equipment-related artefacts;

14.4.1 Non-linear gradient artefact/FOV distortion artefact

The magnetic gradients use in MRI aren´t perfectly linear. At both edges, every gradient starts to curve (Figure 59).[4, 5]

The result of this imperfection is a physical distortion of the body structures, at the edges of the FOV (assuming the FOV is large enough). [4, 5]

The severity this artefact is dependent on the FOV used, the quality of the gradients, B_0 and the sequences used. [4, 5]

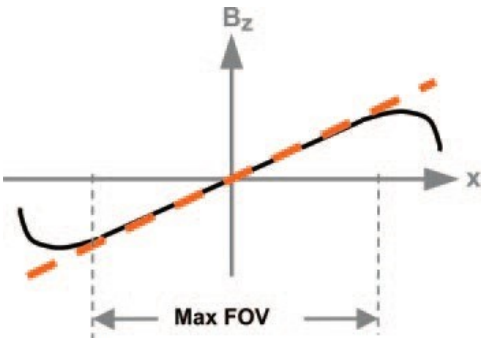


Figure 59: Black line indicates real form of the magnetic gradient. Orange non-continuous line indicates ideal form of the gradient.

It can be mitigated through the use of specific filters, but the only real solution for this artefact is to reduce the FOV. [4, 5]

If the area of interest is very large, multiple sequences with medium or large size FOVs should be acquired and then composed together (e.g. Whole body scan).

SE sequences show good resistance to this artefact, even using the maximum FOV. SE STIR is more sensitive, but large FOV are still partly acceptable.

The same type of testing as the one indicated for Susceptibility artefacts located at the edge of the FOV (end of subchapter 14.3.1) should be performed.

14.4.1.1 Halo artefact

It´s the result of a receiver coil incorrect tuning. Tuning is done in the pre-scan phase of sequence acquisition, after the system determines and sets the receiver gains of the coil and before starting running the sequence. It´s occurs more often when the receiver gains are set manually and appears as increased brightness around the object.[5, 7, 11]

It can only be corrected by proper tuning of the coil.[7]

14.4.1.2 Zipper artefact

This artefact appears as an entire Phase line of alternating light and dark pixels. Sometimes the thickness of the zipper artefact can include several adjacent Phase lines.[5]

It´s caused by external RF radiation being picked up by the receiving coils. It´s most often associated with faulty cables, plugs or wiring in the receiving coil. It can also be caused by the room shielding not being good enough to shield the scanner from electromagnetic waves originating from outside the scanner room.

Careful assessment about the origin of the artefact needs to be carried in order to sort the problem. In the best case scenario is the result of improperly connecting one of the coil's plugs.

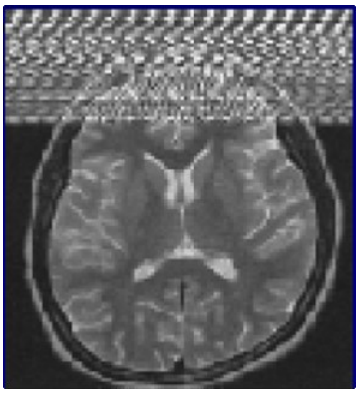


Figure 60: Axial T2 of brain. Zipper artefact is seen in multiple lines at the anterior aspect of brain.

14.5 Environment-related artefacts

14.5.1 Corduroy/Herring-bone artefact

Just like the Zipper artefact, Corduroy is caused by interference of external RF radiation with the MR system. [11]

It is normally contained within the group of images that were being acquired when the interference occurred. It appears as alternating low-high intensity diagonal stripes.[11]

The best course of action is to repeat the sequence and check for potential causes outside the scanner room. [11]

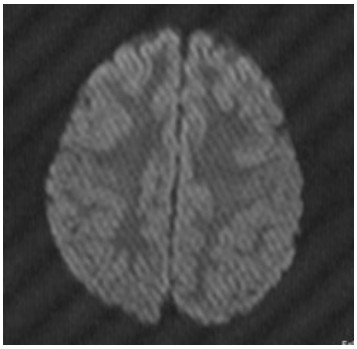


Figure 61: Axial Flair of brain. Corduroy artefact is seen throughout the entire FOV.

It's unlikely that it would occur systematically, but in such scenario it will probably be caused by electronic equipment recently installed, repaired or stored near the scanner room.

14.5.2 Spike artefact

Spike artefact is visually similar to Corduroy artefact but it's caused by the occurrence of a noise spike within the Raw Data. [11] It´s normally a spontaneous non-repeatable occurrence, but when appears often a service maintenance of the MRI scanner should be performed.

It can be corrected by removing the noise spike from the Raw Data. [11]

The easiest way to solve the problem is by repeating the sequence. [11]

15 Considerations for 3 Tesla MRI

The maximum allowed SAR can be more easily reached at 3T (and higher field strengths) than in lower strength magnets due to the increased radiofrequency deposition that results from using higher frequency RF pulses and gradients with higher amplitudes and Slew Rates.[5, 7, 11, 13]

T1 relaxation times are longer than at lower field strengths, leading to reduction of T1 contrast (for the same TR) in multi-slice SE.[5] T1 contrast can be improved by reducing the excitation flip angle (to around 70º) and/or increasing TR (around 800ms). Alternatively, IR-TSE sequences can also be used to create T1 images (TR needs to be at least 2 times TI (e.g. TR: 2000ms, TI: 800ms)).[5]

T2 relaxation times are slightly shorter at 3T. As a result, shorter TE's can be employed without T2 Weight detriment. GRASE is a good alternative to TSE due to low SAR (from less 180º pulses), although contrast between tissues is reduced. Fat suppression is important in GRASE because Chemical Shift artefact is more significant than in TSE.[5]

SNR is significantly improved at higher field strengths.[5, 7, 11, 13, 23, 29] Theoretically SNR improves in the same order of magnitude as the field strength. In practice, due to technical limitations and increase in the significance of other detrimental factors (e.g. Noise in the wires, need of higher BW), the increase in SNR is only on the magnitude of around 41% when B_0 is doubled from 1,5T to 3T.[7]

Significantly higher values of rBW need to be employed compared to lower field strengths, to obtain similar Chemical Shift per pixel.

Overall, most artefacts tend to become more prevalent at higher field strengths.[7, 11] Susceptibility artefacts are enhanced drastically, to a point where MARS can present disappointing results or even inability from the scanner to pick up enough SNR to tune the receiving coils. This makes the scan of areas with large metal implants highly unadvised.

3D scanning is much more feasible at higher 3T and higher field strengths, due to reduced scanning time and improved resolution, without compromising SNR (because it's intrinsically higher).[5, 7, 11, 13, 29]

Section 3: Methods

16 Equipment, materials and human support required

The MRI equipment used were a Siemens Symphony (1.5 Tesla) MRI scanner, with Total Image Matrix (TIM) technology, a Philips Achieva X series (3 Tesla) and all their respective coils available. All these equipment belongs to the University of Oxford NHS Trust and is currently located and in clinical use at the location where the project was developed.

Additionally Quality Assessment (QA) MRI Phantoms and a biological phantom were used during the whole process of sequence optimization. The QA phantoms were used to evaluate the SNR, the presence of artefacts and coil working conditions, while the biological phantom was used in order to assess all aspects of the image quality and weighting.

Some human volunteers were used to test the final sequences created by the author and to obtain the MRI images that were used in the image assessment part of this project.

The project required only a single radiographer (the author of the dissertation) to manipulate the sequences. The assessment of the final sequences' image quality was done (upon request by the author of the dissertation) by fellow MRI radiographers and Radiologists of the NOC.

17 Phantom testing conditions

17.1 Scanning Conditions

All scanning undertaken for optimization of sequences was performed using non-living non-human soft tissue phantoms.

Two MRI scanners were used:

1. 1.5T Siemens Symphony a TIM system
2. 3T Philips Achieva X series

No other special conditions were employed.

All available receiving coils of both MRI systems were used for scanning the phantoms.

Post optimization, the sequences were applied to the clinical setting and only after the resulting images were deemed has having adequate diagnostic quality, by the scanning radiographer, were they anonymized and processed to be included in the image evaluation method. Additionally clinical protocols were created with these sequences for possible future application.

18 Optimization Methodology

Initially the author tried to establish the ideal Signal-to-Noise Ratio (SNR) and Contrast-to-Noise Ratio (CNR) for all MRI image weightings, used at the NOC. This was done by scanning biological and non-biological phantoms, on a try-and-error base method, using different combinations between Number of Excitations (NEX), Field of View (FOV), Oversampling, Slice thickness (3 mm and 5 mm), Matrixes, Bandwidth, MRI coils, etc.

The combinations were done for the following types of MRI Weights: T1, T1_Fat Sat, T2, T2*, T2_Fat Sat, PD, PD_Fat Sat and STIR.

After the ideal combinations, to obtain a good enough SNR and CNR were found (Figure 62) for each type of Weight, twelve combinations, for both slice thicknesses used, were selected to be used as sketches for the development of Template Sequences":

- 1st Sketch:
  - Low Resolution (≈1.1mmX1.1mm)
  - Large size FOV
  - Medium size FOV
- 2nd Sketch:
  - Medium Resolution (≈0.6mmX0.6mm)
  - Medium size FOV
  - Small size FOV
- 3rd Sketch:
  - High Resolution (≤0.4mmX0.4mm)
  - Medium size FOV
  - Small size FOV

FOV size was categorized as following:

- Large FOV:
  - Between 300mm and 500mm;
- Medium FOV:
  - Between 200mm and 300mm;
- Small FOV:
  - Between 50 mm and 200mm.

The Sketch sequences were chosen considering the best balance between acquisition time, image resolution, maximum number of images per TR and Turbo Factor/Echo Train. The ruling as the following:

- Lower the acquisition time and Turbo Factor/Echo Train was preferred;
- Higher image resolution and maximum number of images per TR was preferred

Template Sequences were built from the Sketch Sequences chosen. The Templates had 30 slices each and the parameters that don't affect SNR were adapted in order to keep acquisition time as low as possible, while maintain TR and TE within image type range. Turbo factor was kept as low as possible to avoid image blurring. Whenever parallel imaging was used, NEX, Oversampling and other techniques were changed to correct for SNR decrease.

Template-sequences were then tested and only after satisfactory results, adapted to specific examinations and specific coils.

The optimized sequences were tested on a biological phantom and after this step was concluded satisfactorily, they were finally compiled into clinical MRI protocols.

The method used to assess the image quality of the optimized sequences was visual evaluation by Musculoskeletal/Orthopaedic Radiologists and MRI Radiographers. Radiographers were enquired about the technical quality of the images, while Radiologists were asked to comment on the images' technical quality and diagnostic quality.

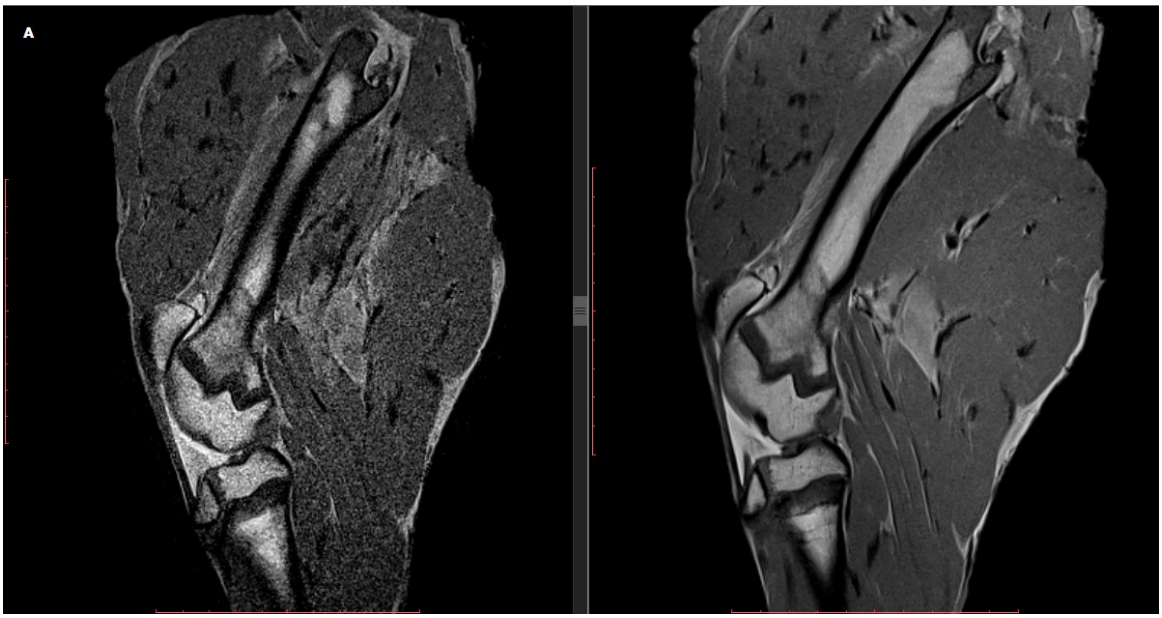


Figure 62: Sagittal T1 TSE of biological phantom used. Image A was obtained with a basic non-optimized sequence while B was acquired after the sequence, used to get image A, was optimized for SNR and CNR.

19 Optimizing sequences

Just like baking a good cake, optimizing sequences is all about finding the best relation/balance between all the ingredients (parameters) available. Having an idea of what your final product should look like or what its defining feature should be is fundamental for success. After all, baking a tart is different from baking bread and baking a regular lemon tart is a bit different from baking a low calorie lemon tart. The same thing happens with MRI sequences. Knowing which main aspect(s) (e.g. Contrast, TA, SNR, Spatial Resolution or Artefacts) of the sequence need(s) improvement is essential. In this project the author aimed at reducing both TA and Artefacts and at improving Spatial Resolution.

After deciding what the defining feature(s) of the final product will be, the only thing that is left to do is to know how to get there. Essentially, after choosing what type of cake to bake and gathering all the ingredients needed, you have to get the cake's recipe. In MRI, this means that:

1. One must understand how TA and Image Quality factors relate to each other (how the main ingredients interact between them) (see Chapter 20);
2. Define and follow a methodology that takes in consideration the known variables of the process, while leading you to the final product (how and when to mix the ingredients and what baking tools are necessary) (see subchapters 19.4).

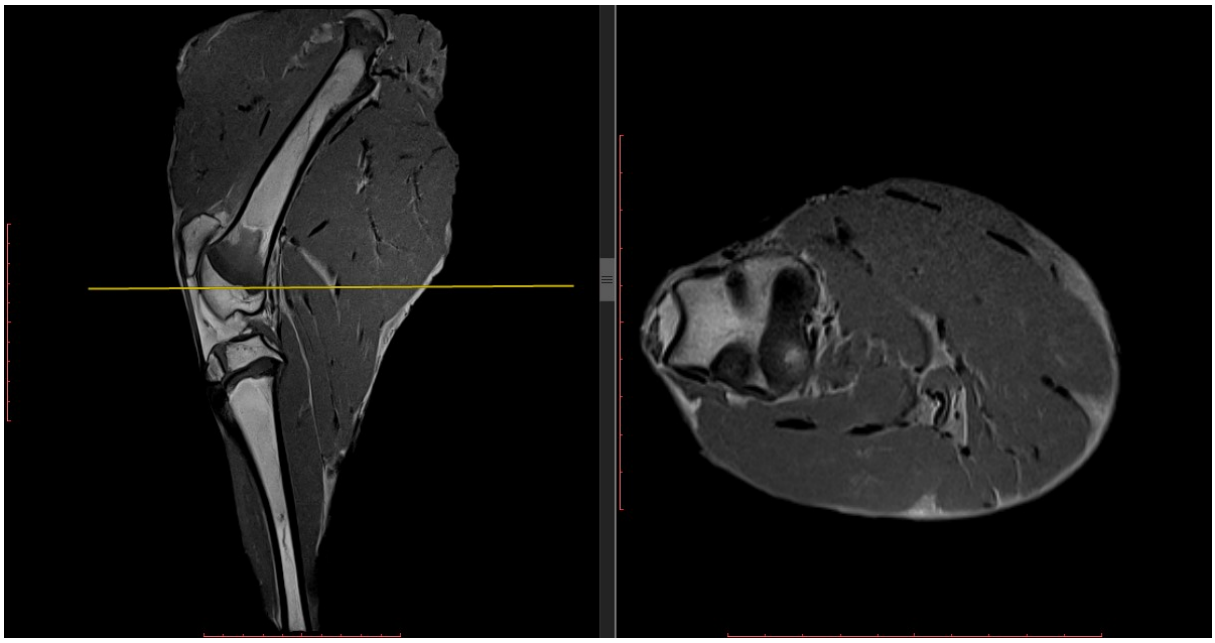


Figure 63: T1 TSE Sagittal and T1 TSE Axial planes of biological phantom. Large & Medium FOV template sequences.

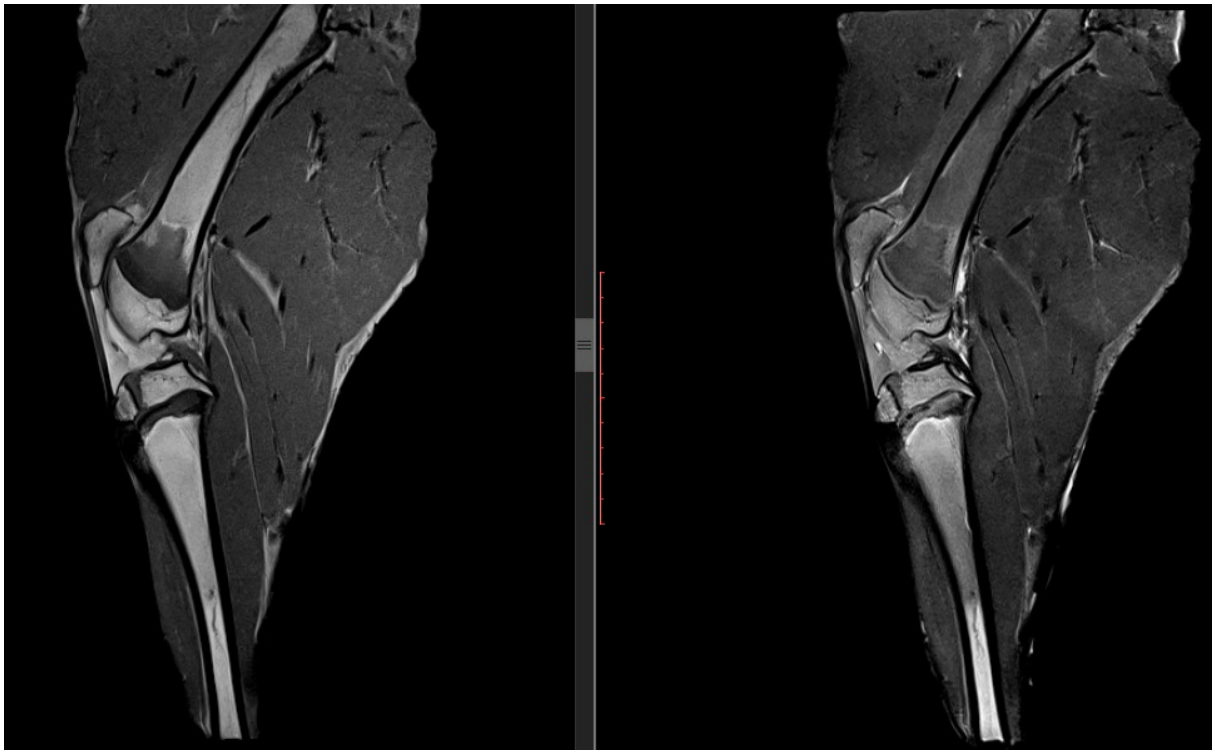


Figure 64: Sagittal T1 &T2 TSE of biological phantom. Large FOV template sequence.

19.1 Selecting a Phantom

The first point to consider when optimising sequences is what phantom to use. Ideally, using a human test subject would be the best choice, but considering the time it takes to develop optimized sequences and how boring the process is for the test subject(s), it is difficult to find someone who is willing to spend so much of their time and patience inside a MRI's bore.

Water or oil Phantoms are good to assess the presence of acquisition-related artefacts, equipment faults and low SNR but they provide little or no insight regarding image Weight and

Contrast. They are best suited to test for the presence of acquisition-related artefacts after the sequence has been optimized for the SNR-TA relation.

Biological phantoms (Figure 65) made of meat (e.g. Leg of lamb) are the best option because their multi-tissue structure is more similar to clinical images in both contrast and anatomy. In addition, their size and shape are more adjusted to human body parts, than water or oil-based Phantoms, enabling the conditions under which sequence optimization is undertaken to be the exactly the same as when scanning human subjects. Nevertheless, issues regarding hygiene and infection control must be considered when using this type of phantom in clinical scanners and so safety measures must be defined and put it place (see pages 140-144).

For this project a lamb leg was used (Figure 65). Its conical shape enables the simulation of both small and medium size both structures, plus it adapts well to most coils.

It was noticed during the optimization process that, although the image's Contrast and Weight was similar to clinical images, the signal intensity of marrow tissue and other fatty tissues wasn't exactly the same as in clinical images of human subjects. It appeared significantly more hypointense intense than expected (Figure 66 & Figure 67). This was likely partly due to the young age that the lamb possessed, since similar loss in signal intensity in these tissues is also observable in human infants, and partly because the phantom is not living tissue.

Figure 65: Example of possible Biological MRI Phantom. In this project a vacuum sealed whole leg of lamb, identical to the one shown in this image, was used (approx. 1.4 Kgs in weight). It is widely available for purchase in any large comercial food retailer already packaged encased in vaccum inside a thick/resistant plastic cover. This form of encasing ensures that the meat remains fresh for longer and minimizes the risk of contaminating the MRI equipment with potential harmful bacteria.

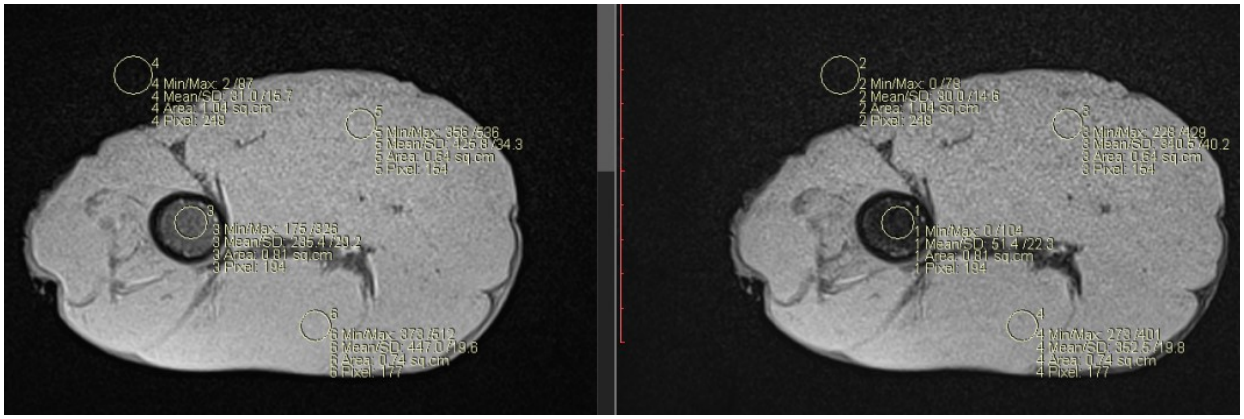


Figure 66: Axial view of a Lamb leg. Right image: T1 TSE. Left image: T1 with full Fat saturation. The same as in Figure 67 is seen. Additionally, significant Chemical Shift artefact reduction is seen when Fat saturation is employed, as well as overall SNR drop.

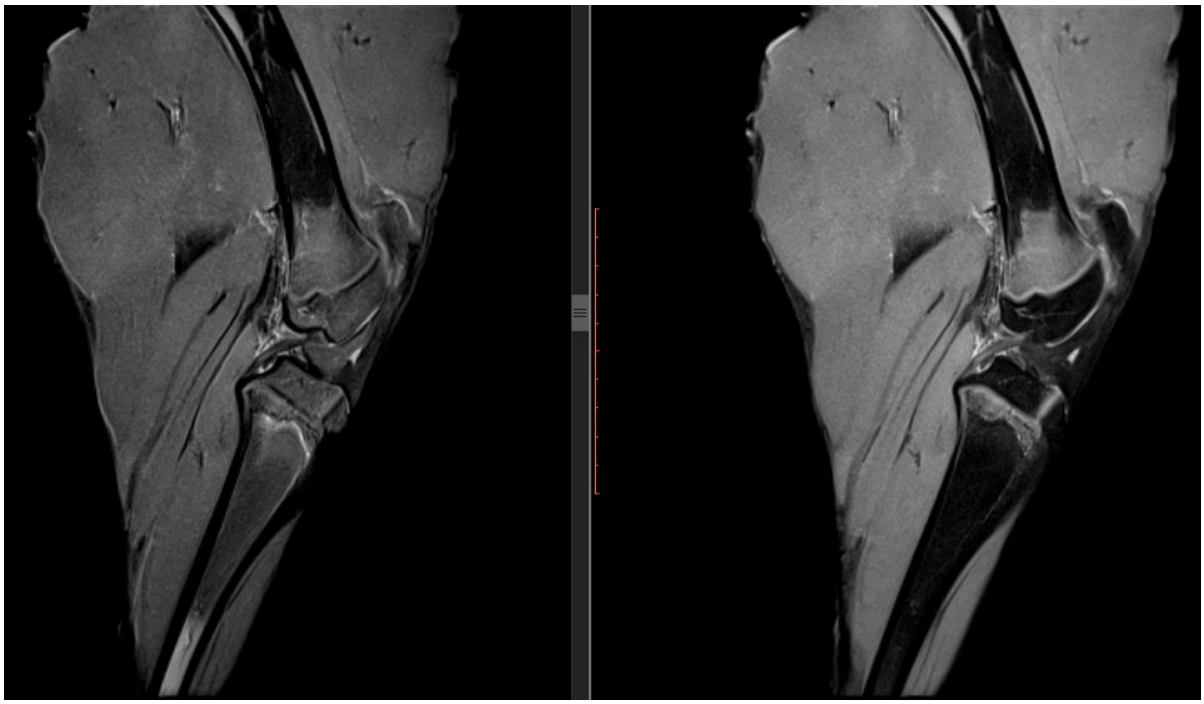


Figure 67: Sag view of a Lamb leg. Right image: TSE PD. Left image: TSE PD with full Fat saturation. Although tissue saturation is obvious, the signal intensity of the bone marrow on the image on the right is still fairly low considering that it's PD Weight. TR: 2750ms; TE: 24ms; B0:1.5T; refocusing FAng: 180º.

19.2 Finding how much SNR is needed

The most common question that a Radiographer/MRI operator faces when adapting a sequence to the scan been performed is simply: "Is there enough SNR?"

The answer is too often: "I don't know".

Literature [7, 11, 40] tells us that a SNR (Equation 16) higher than 20:1 is unlikely to produce any benefit to image quality from an observer's point-of-view. One could then assume that 20:1 would be the best SNR to have, but if TA is considered it's likely that a lower ratio would be preferable has the relation between SNR and TA is logarithmic (hence why SNR over 20:1 is not interesting) and not linear.

$$SNR = B_0 * K * Voxel\ Dimension * \sqrt{\frac{number\ of\ measurements}{Receiving\ Bandwidth}}$$

Equation 15: Basic SNR calculation formula for MRI. Adapted from [7, 11, 40]

K = numerical constant that encompasses the influence of the MRI environment: distance between FOV and coil(s), coil(s) used, other hardware dependent factors (gradients, RF transmitter, etc.), pre-amplifier, noise power spectrum, field strength dependent factors (B_0 homogeneity, shimming's effective volume, distance between FOV and B_0 centre, etc.), sequence type, sequence parameters (TR, TE, FAng(s), TF, etc.), tissue dependent parameters (T1, T2, PD), quality of the room's electromagnetic insulation, external influences, presence of objects with susceptibility properties, age of the MRI scanner, scanner's temperature, etc.

$$Voxel\ Dimension = \frac{FOV_{FE}}{N_{FE}} * \frac{FOV_{PE}}{N_{PE}} * effective\ slice\ thickness$$

$$Number\ of\ measurements = N_{FE} * N_{PE} * Number\ of\ excitations$$

$$SNR = B_0 * K * \left(\frac{FOV_{FE}}{N_{FE}} * \frac{FOV_{PE}}{N_{PE}} * ST\right) * \sqrt{\frac{N_{FE} * N_{PE} * NEX}{rBW}}$$

Equation 16: Advanced SNR calculation for MRI. Adapted from [7, 11, 40]

Despite all that was stated in the previous paragraph, since SNR is affected by so many factors and a significant amount of them are, in practice, unknown variables (K in Equation 16) within the equation for SNR (Equation 16), it becomes practically impossible to use objective numerical values for SNR in day-to-day clinical scanning. It can be calculated post-acquisition, by measuring the signal intensity distribution within the images, but it is unknown during the planning phase of the scans. This is why manufacturers refrain from using SNR as a parameter and use instead relative SNR (rSNR). This parameter doesn't provide a real value for SNR. Alternatively, it shows the mathematical ratio between the theoretical SNR of the base sequence and the theoretical SNR of base sequence plus the parameters changes applied at the moment of sequence planning. The base sequence is assumed to always have rSNR equal to 100% (or 1, depending on the manufacturer). If the changes made are saved then the rSNR will be set as the new 100% (or 1), regardless of the actual rSNR value.

The use of rSNR allows for a comparison between sequences, but implies that the base sequence has, itself, enough/adequate SNR. This may well not be the case, either because the scanning conditions (MRI's environment) have changed or the base sequence was itself incorrectly set up. Additionally it's quite difficult to correlate between an objective numeral value in rSNR and the subjective perception of noise in an image.

For these reasons finding the right SNR in clinical day-to-day MRI scans is normally done through trial-and-error and the radiographer's personal experience.

The best approach is to design several template sequences/protocols with several combinations of case non-variable parameters, which provide adequate SNR (and not excessive SNR) for a variety of different cases (e.g. normal patient/obese patient; dedicated coil/general coil) (see subchapters 19.3; 19.4; 19.5). This can only be achieved through trial-and-error methodology.

As one of the fundamental factors of Image Quality (IQ), SNR needs to always be kept under careful scrutiny by the operator:

- Low SNR will display as a speckle/grainy look on MRI images, that can extend through the entire FOV or be located only at the centre/edge of the FOV. Images with non-homogeneous grainy look are the result of there being a significant distance between the receiving coil and the area where the grains appear. Placing an additional receiving coil near such areas will correct the problem, yet that is not possible or advisable for all situations (e.g. Thoracic spine exam). In such cases increasing rSNR

(and subsequently SNR) by parameter manipulation is the only solution, even though it can cause a significant increase in TA or decrease in spatial resolution. Additionally, low SNR will affect contrast negatively, through CNR. It is then of the upmost importance to avoid low SNR, when setting up any sequence.

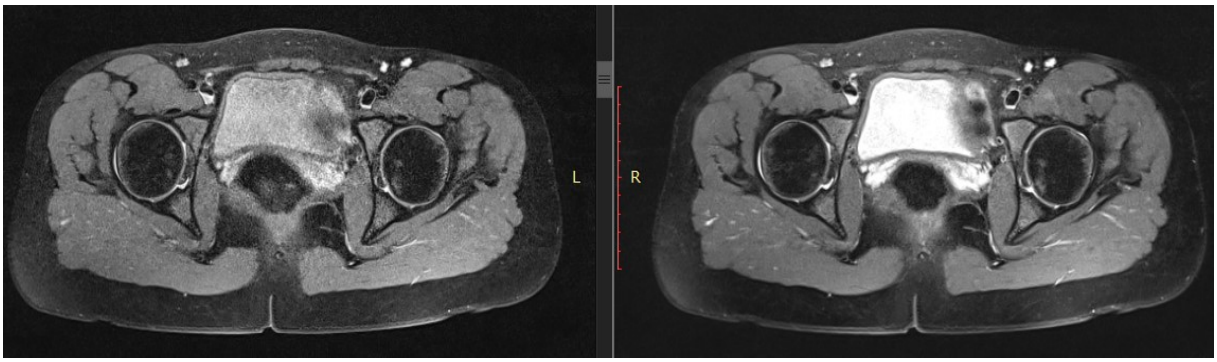


Figure 68: Axial PD_FS TSE of pelvis with full Fat saturation. The image of the right exposes low SNR (and subsequently low CNR). TA: 2min 24 seconds. Image on the left acquired with additional SNR. TA: 3 min 11 seconds. Slice thickness: 3mm; TE: 25ms; Number of slices: 34.

- Excessive SNR will make images appear slightly blurred or foggy, due to tiny twitches of motion that most patients experience over time and get registered when the sequence's TA is long. The image's CNR will be partially improved but it implies that a significant amount of time is being wasted and there is increased risk of significant motion artefacts on the images of that sequence and any sequence after that (since most patients struggle to keep still for long periods of time).

19.3 Adapting sequences to the coil(s) used

Defining à priori the MRI environment/conditions under which optimization is performed is fundamental for a proper relation between the results obtained during optimization and the results obtained with real human patients. While most MRI environment-related factors are constant for the scanner used (e.g. Field's strength) there are some that can change depending on the body part scanned (e.g. Position of the scanned structure inside the bore). Among these, the receiving coil used is one of the most important factor and unfortunately the operator often forgets to take into account (while scanning) that not all coils are the same (see subchapter 6.5.4).

Several coil properties affect the resulting SNR:

- Number of channels;
- Type;
- Size;
- Shape;
- Being transmitter/receiver or just receiver;
- Etc.

In order to tackle these differences, sequence optimization must be performed for all coils.

This doesn't mean that one must start from scratch with every coil(s). It just means that after optimization is performed in one coil (preferably the one most routinely used in that scanner), the optimized sequences are also run with the rest of the coils (in the most usual scenario under each coil(s) is used) and, if necessary, adjustments are applied (e.g. increase or reduce SNR accordingly). In this project, optimization was first performed for a combination of Spine and Body coil (for the 1,5T scanner), and Knee Coil (for the 3T scanner).

It's fundamental to label the optimized sequence for their respective coil to avoid future mix-ups.

The same principles apply to different image's Weights, FOVs and spatial resolution.

It's also important to take notice if IPAT is used, since some coils and/or some scanning scenarios do not allow for the use of parallel imaging.

19.4 Developing sequences for Small, Medium and Large FOV

As indicated in the beginning of this section, several template sequences with different FOVs and Spatial Resolution were produced and optimized for each Weight and receiving coil(s) at both 1.5T and 3T.

The sequences were labelled accordingly for:

1. MARS or non-MARS;
2. Weight;
3. With or without Fat saturation;
4. Type of sequence;
5. Acquisition plane;
6. FOV in the frequency dimension;
7. Matrix/pixel resolution;
8. rBW;
9. Use of IPAT (and respective AF);
10. Coil(s) used;
11. Presence of post-acquisition filters.

The process was performed by starting with the defined parameters indicated in Chapter 18 and then by manipulating the MRI parameters according to the knowledge, exposed in this Dissertation, and the Author's best judgment.

With the aim of achieving the best SNR-TA relation, a group of three different Spatial Resolutions (considered either adequate or more than adequate for clinical MRI) were defined à priori. A trial-and-error methodology was then applied, where the Author deliberately increased or reduced the sequence's SNR (through manipulation of multiple MR parameters) until images with satisfactory IQ were obtained using the minimum SNR and lowest TA possible.

Satisfactory IQ was defined visually and taking in consideration the following factors:

- Weight;
- Contrast;
- Severity of Image's graininess/speckling;
- Presence and severity of artefacts
- Overall visual aspect/appeal of the images.

This process of SNR-TA optimization was performed for all MRI Weights used in MSK.

19.5 Adapting sequences to body part

Adaption of sequences to body part was done by choosing the optimized sequences with the correct FOV and Spatial Resolution for each specific body part.

After this, the necessary MR parameters changes were introduced (while maintaining the same SNR) in order to adapt the optimized sequences the peculiarities of each body part.

The changes introduced aimed at preventing/attenuating the occurrence of artefacts, avoiding IQ degradation, adjustments for the desired plane(s) of acquisition, and correcting for idiosyncrasies related to the area of the body in question (e.g. breath-hold acquisition in sequences for the ribs.)

The final sequences obtained were grouped together according to the NOC's MRI Protocols. This resulted in the Optimized MRI Protocols.

The images produced by these protocols where later assessed by both NOC Radiologists and NOC MRI Radiographers, according to the methodology disclosed in Chapter 22.

The results of the image assessment are exposed in subchapter 23.1.

20 Time of Acquisition vs. Image Quality

All MRI scanners have limitations on the quality of the images they can produce. Even though achieving this limit is the ultimate aim, the reality is that such limit is very rarely (if ever) achieved because of another much more common type of limitation: TA.

High IQ and short TAs are both highly desirable characteristics of any MRI sequence and it's not sensible to overlook any of the two.

The aim must then change from: Achieving the best IQ possible; to become: Obtaining the best IQ within the shortest TA possible. This new aim can be referred as Optimal Image Quality and it's particularly hard to achieve due to how TA and IQ relate to each other.

TA is a very objective concept that can easily be measured, while IQ is the exact opposite. They relate to each other by three very important IQ factors: Spatial Resolution, SNR and the presence of Artefacts. As exposed in previous chapters, improving any of these three factors demands (usually) an increase in TA. As a result of these trades-offs, a balance needs to be ascertained in order to achieve Optimal Image Quality (Figure 69 & Figure 70).

To understand how to obtain Optimal Image Quality it is important to bear in mind four things:

- SNR and Spatial Resolution are more desirable the higher they are;
- TA and Artefacts are more desirable the smaller they are;
- Contrast (and by extension also Weight) can also influence IQ, although not always in a linear way.
- Why and how changing one factor will also likely change the others.

Quite often, trying to bring down TA while keeping IQ in check requires tweaking multiple MR parameters (considering the size of the patient, the body part scanned, its proximity to the centre of the bore, the coils used, SAR, etc.) so the overall trade-offs between parameters will result in a good balance between TA and IQ (Figure 71).

Experience in doing the type of examination required and knowing the limitations of the scanner are the essential when trying to achieve the best balance.

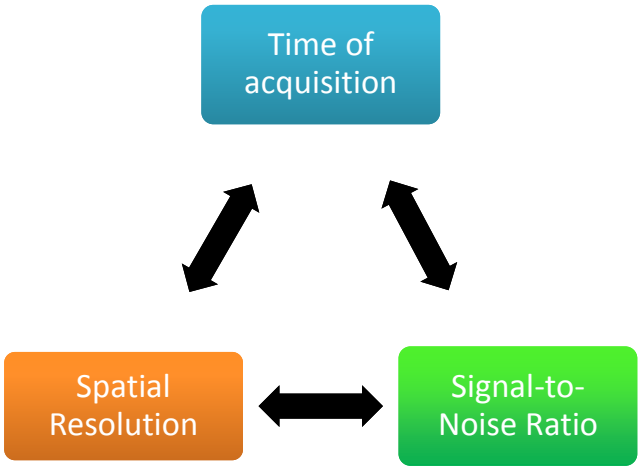


Figure 69: Simplified relational diagram of MRI's main factors at play for Image Quality. The so called: Bermuda Triange of MRI. The arrowss indicate trades-offs between parameters.

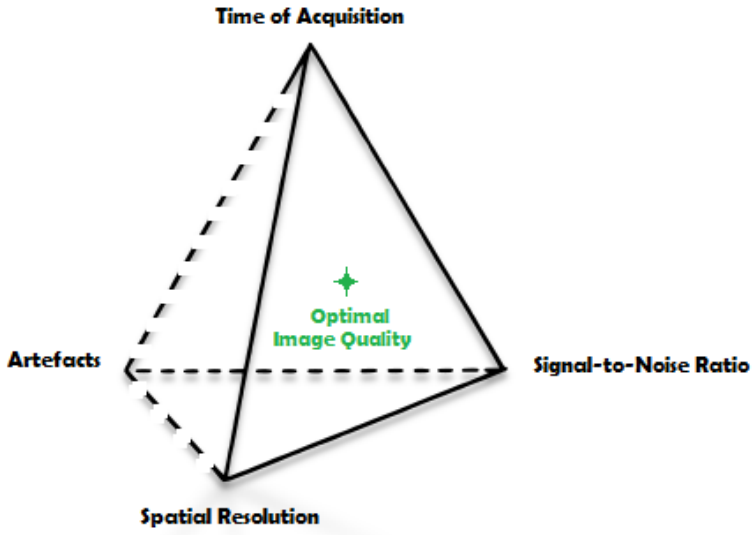


Figure 70: Relational diagram of MRI's main factors for Image Quality. Continuous lines of the triangular pyramid indicate assured trade-offs between parameters. Non-continuous lines indicate potential trade-offs depending on the case at hand. The presence and severity of Artefacts is something that always needs to be considered to attain Optimal Image Quality. This factor is often overlooked when discussing the relation between parameters because of its non-linearity.

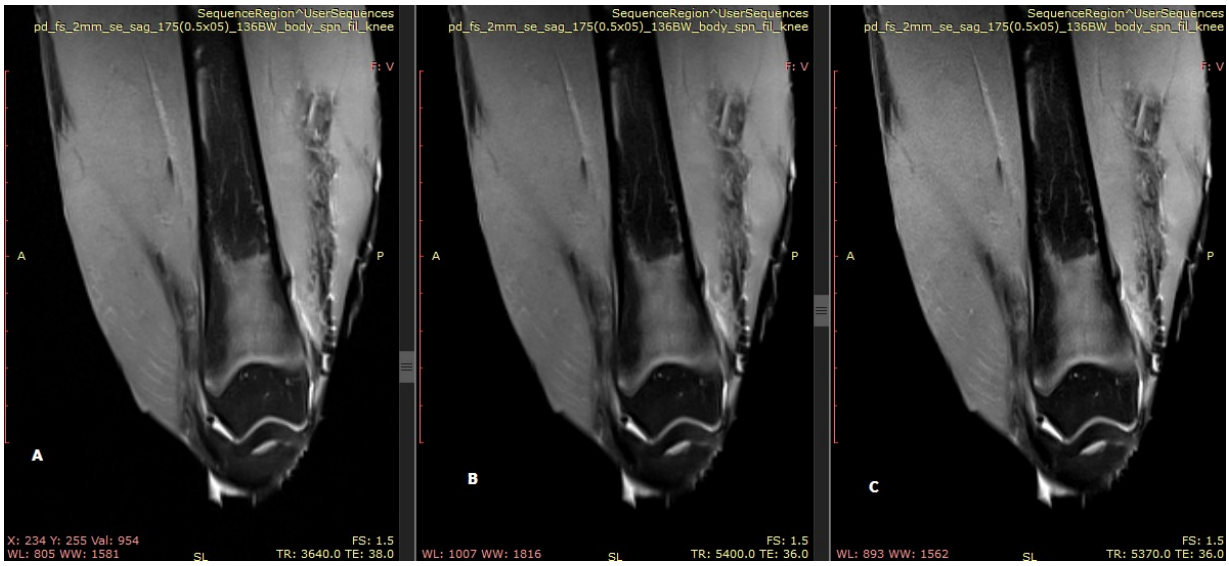


Figure 71: Coronal PD_FS TSE of biological phantom. Images A to C show the results of manipulating key MRI parameters (TR, TE, saturation pre-pulse, etc.) in order to improve the degree of fat saturation and image's sharpness and obtained and optimized CNR (C). The TA was similar (less than 10 seconds of difference) for A, B &C.

20.1.1 TA vs. Spatial Resolution

Increasing the image's in-plane Spatial Resolution is done by increasing the image's matrix (or reducing the FOV). This normally (but not always) implies an increase of both N_{FE} and in N_{PE}. While N_{FE} is almost unrelated to TA (only through the Slew Time of gradients), N_{PE} is. As exposed previously in subchapter 9.1.4.2, TA is highly dependent on N_{PE} because with each extra N_{PE} added to the matrix, an extra TR is needed in order to fill it. The relation is pretty much linear. The best course of action, for a low TA, is to keep N_{FE} high and N_{PE} low (around 65%-100% of N_{FE}). It's important to avoid pixels with a very rectangular shape (especially when the pixel's size is big), so the lower the N_{FE} is the closer to 100% N_{PE} should be.

FOV also has some influence in the Spatial Resolution–TA relation. When FOV is low, small changes in the N_{PE} will have greater impact in varying the Pixel's Phase dimension, than at medium or large FOVs. The opposite is true for N_{FE}. In situations where the FOV is low and the Base Matrix is already significantly high, improving the N_{FE} to the next available higher value won't result in any substantial improvement in the Pixel's Frequency dimension (but it may in TA). In this type of scenario is best to choose the lowest N_{FE} value that provides the desired pixel dimension.

When optimizing for the TA-Spatial Resolution relation, is best to start by having a notion of what dimensions the pixels should have and then:

1. Define the slice thickness as desired for the protocol;
2. Define the lowest FOV allowed by the scanned structure(s)/area(s) of interest;
3. Increase the Base Matrix until the Pixel's Frequency dimension equals (or is very close to) the desired resolution;
4. Gently increase the size of the FOV until the point where the Pixel's Frequency's dimension is about to change;
5. Increase N_{PE} just until the desired pixel dimension is achieved;

6. Adjust the SNR until the Relational SNR value is at value 1.00 (or 100%, depending on the scanner) by changing any parameter that affect SNR but not the Resolution;
7. Improve both RF pulse and gradient mode to the maximum level (or normal level if SAR is a concern);
8. Check TR and adjust it accordingly to the Minimum TR;
9. Check the TF value and increase it to the point where ET is long but not too long (see subchapters 9.2.1 & 9.2.2) to produce significant blurring (since that would degrade the image's resolution and sharpness);
10. Re-check TR, adjust TE and change Concatenations' value accordingly until the minimum TR is within range for the desired Weight;
11. Re-check TR and adjust it accordingly to the Minimum TR.

After all these steps the sequence will be optimized for TA-Spatial Resolution relation. If Interpolation is used, then divide the size of both N_{FE} and N_{PE} by the factor of Interpolation (which shouldn't be higher than 2).

20.1.2 TA vs. SNR

The aim in the TA-SNR relation is to find out the minimum SNR needed for the defined Spatial Resolution and the Patient at hand (bigger patients require more SNR because of the distance between receiving coil and entire FOV).

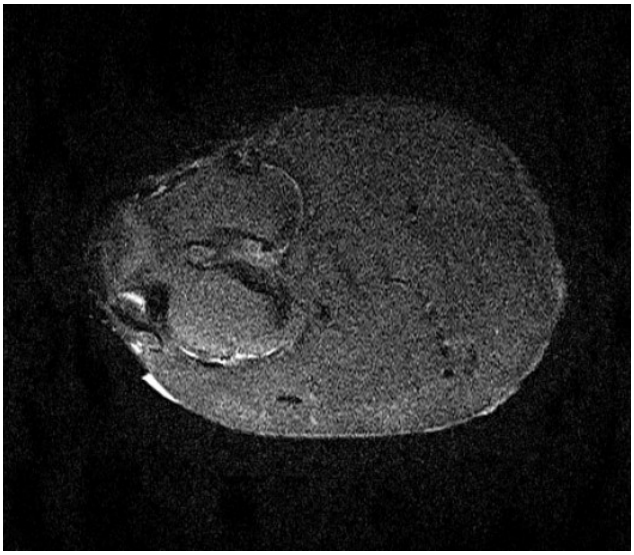


Figure 72: Axial image of biological phantom obtained from a sketch sequence. Classic example of how extremely low SNR leads to a complete compromise of all aspects of IQ. The high level of noise seen spreading outside the scanned structure tells us that it was the MRI parameters and not the structure that cause the poor IQ.

As discussed in subchapter 9.2.8.2, SNR is dependent on the number of measurements/samples that the system performs. Each measurement requires time (even though an almost insignificant amount of time), so more measurements will add up to more time. Good SNR requires thousands upon thousands of measurements and significant variations in SNR require huge variations of the total amount of measurements performed (normally done by varying NEX or Oversampling). TA varies quite a lot with significant SNR changes (see subchapters 9.1.6 & 20.1.2).

The relation between the two is not linear. Figure 73 exposes the increase in the discrepancy between TA and SNR with the increase of NEX. As a result, any sequence optimized for time will have to keep NEX as low as possible (between 1 and 3 NEX).

In the case of Oversampling, it behaves similarly, but since this parameter is defined as a percentage, its maximal value is 100. When Oversampling changes from 0 to 100%, this represents an increase in SNR of 41%, while TA doubles. Contrary to NEX, Oversampling can be fine-tuned and so it provides more flexibility for adjusting the TA-SNR balance. This doesn't apply to some of the newest MRI scanners, were fine-tuning NEX is also possible to a good degree.

If a large variation in the TA-SNR balance is desired, NEX is a better parameter to be changed, while Oversampling is best for small or medium variations in the balance.

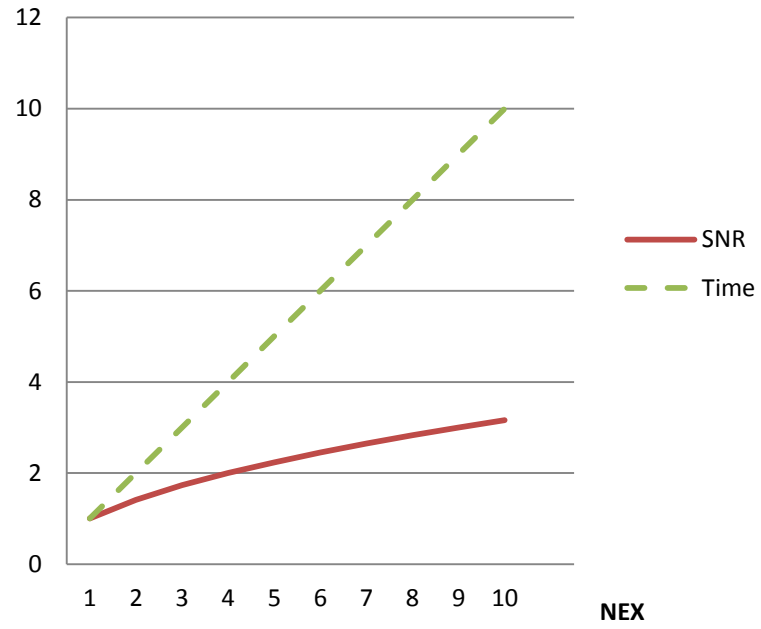


Figure 73: Relation between NEX, SNR and Acquisition Time.

Half Fourier and SENSE can be used in the same way as Oversampling (if scanning conditions allow for such parameters to be employed). The rBW can also be used in this fashion but to a smaller degree due to its indirect non-linear trade-off between SNR and TA (see subchapter 20.1.2). It's acceptable to increase rBW to reduce TA if the decrease in SNR is between 1% and 10%, but more than that and the sequence will not be optimized for time. Attention must also be paid to the change in Chemical Shift per Pixel introduced by the variation of rBW.

Parameters like GRAPPA's AF can be used in the same way as NEX, but in this case higher values of AF mean a decrease in SNR and TA. From a TA-SNR balance point-of-view, NEX and GRAPPA's AF are antithesis of each other and will "nullify" each other's effect if their values are equal (e.g. NEX=2 and GRAPPA's AF=2). For this reason, their values should always mismatch (e.g. NEX=3 and GRAPPA's AF=2), preferably with NEX having the lowest value among the two parameters (for TA's sake).

20.1.3 TA vs. Artefacts

Artefact reduction may or may not affect TA. It all depends on the nature of the artefact and what is done (if necessary) to avoid, correct, or suppress it. The ways of how artefact correction affects TA are never direct, yet the following tendency exist: Artefact reduction is achieved at the cost of increase in TA.

This is particularly obvious when the operator is very proactive at reducing a particular type of artefact (e.g. metallic artefact). The lengthening of TA is normally a side-effect of a necessary increase in:

- Minimum TR (e.g. applying a Sat Band to reduce breathing artefact requires pre-pulse preparation which in turn increases the minimum TR);
- Sampling (e.g. extra sampling partially reduces/obscures fold-over and/or flow artefacts);
- Matrix (e.g. Gibbs artefact is only corrected by improving the matrix in the frequency direction);
- Number of TRs employed (e.g. using a shorter ET to avoid blurring of images).

There are many situations (but not always) where if multiple parameters are "juggled" together properly, it's possible to reduce/nullify some artefacts without compromising TA (or even SNR or Spatial Resolution) (e.g. correcting fold-over artefact by improving Oversampling followed by improvement of RF pulse mode and rBW to reduce SNR and the minimum TR).

There is also the particular case of motion-related artefacts caused by involuntary spasms, cramps, pain or discomfort. For this type of artefacts, their incidence increases exponentially with longer TAs. Staying still for long periods of time is hard for any patient, especially for debilitated patients, hence why longer TAs lead to a higher chance of motion-related artefacts.

20.1.4 SNR vs. Spatial Resolution

As exposed before, higher Spatial Resolution means smaller pixels/voxels which in turn implies a lower amount of protons per voxel (see subchapter 9.1.4 for more details). This results in lower SNR, regardless of the scenario at hand because there is lower signal intensity within each voxel. Finding out the optimal relation between SNR and Spatial Resolution is one of the hardest things to do in clinical MRI, because of two important things:

1. Lack of established guidelines/standards regarding the minimum Spatial Resolution required for each MRI examination;
2. Inability to reliably calculate SNR prior to image acquisition.

Experience with the scanner and with the coils used is fundamental in establishing the best balance in the SNR-Spatial Resolution relation.

As it was done in the practical component of this Dissertation, is best to first establish an average scanning environment (including where, how and which coils are used), define the desired Spatial Resolution and then run the same sequence a number of times. Each time the sequence is run a gradual variation in the rSNR should be introduced until the desired image quality is achieved.

If the size of the patient is a relevant factor for the examination, the SNR may have to be slightly adjusted every time the examination is performed. Experience is also important is this

situations, but generally speaking an increase of 30% in the rSNR (at the cost of Spatial Resolution) should provide adequate SNR for large patients.

It is important to bear in mind that graininess from low SNR becomes visually a little less relevant at higher in-plane resolutions but that, in this conditions, Contrast will be partially degraded (due to poorer CNR).

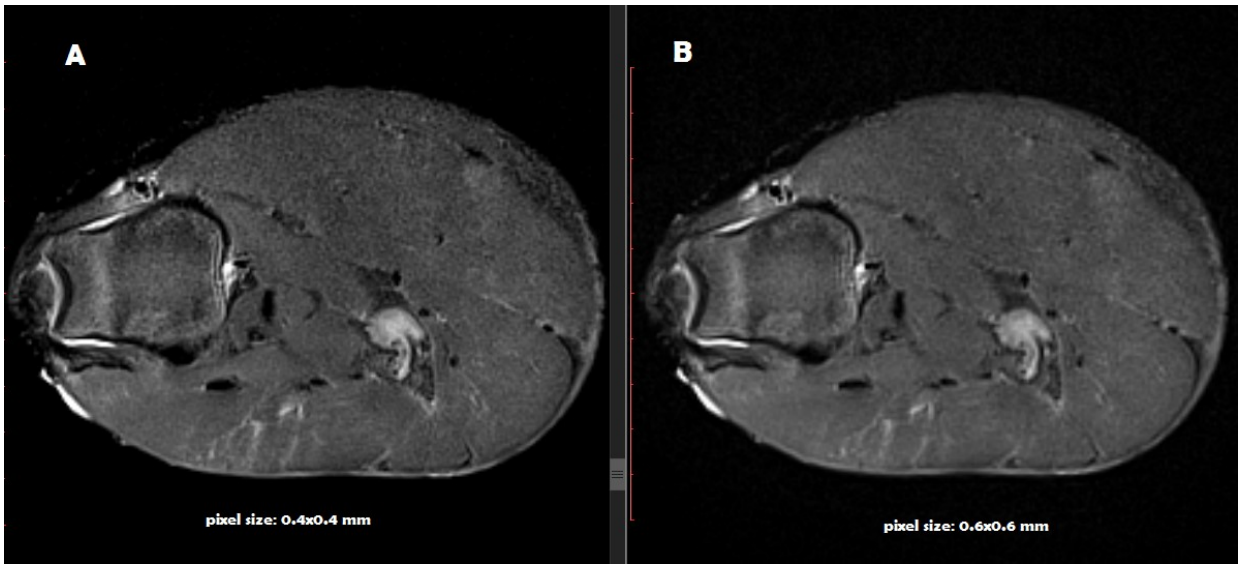


Figure 74: Axial view of biological phantom used (leg of lamb) during sequence optimization. All scanning parameters between A and B (including coils) were similar with exception of the image matrix used (576x576 for A & 384x384 for B), leading to a change in pixel size (0.4x0.4mm for A and 0.6x0.6 for B). Image A shows improved detailed in comparison with B (due to having a smaller pixel size), however the SNR in A is significantly poor to likely compromise the diagnostic quality of the image. Image B shows acceptable SNR, even though some noise is still perceivable throughout the image, When high value matrix are used, only a substantial change in pixel size (about 50% is this case) is able to produce a perceivable improvement in SNR (moving from A to B without increasing TA).

20.1.5 Artefact vs. SNR

As a general rule, Artefacts do not relate to SNR but there are cases were high SNR can correct or obscure some artefacts:

- Images with a grainy visual look are obviously the result of poor SNR. This is an artefact and can only be corrected by higher SNR. It can also be slightly mitigated by a post-acquisition smoothing filter, although image sharpness may be affected.
- Flow-related artefacts are not corrected by changing SNR, yet a larger number of samples will mitigate they severity. When the signal intensity variation that is characteristic of this type of artefact is sampled multiple times, the different intensity values registered, by the multiple measurements, are averaged into a single value and the intensity variation is therefore reduced. The same is also true for truncation-related artefacts.

Active suppression of flow-related artefacts is preferable to simple SNR increase, but since many of MRI parameters that attenuate flow require a large number of measurements to be effective SNR should be kept high.

20.1.6 Artefacts vs. Spatial Resolution

As a rule, Artefacts will either reduce or not change at all with the increase in Spatial Resolution. Once again it all boils down to the nature of the artefact:

- Aliasing-related artefacts are less likely to occur with higher in-plane resolution due to better sampling of the frequency spectrum that composes the analogue signal produced by protons (e.g. Gibbs);
- Partial volume artefacts are less likely to occur with higher through-plane resolution (thinner slices), because of less overlap between the slice profiles of adjacent slices. There are other ways to correct this without changing the Slice Thickness (e.g. increased Slice Gap or interleaved acquisition);
- Improper shimming or truncation-related artefact are don't vary with changed in the Spatial Resolution;
- Susceptibility-related artefacts are reduced at higher resolution (both in-plane and through-plane);
- Flow-related artefact will be enhanced and blurriness reduced with higher Spatial Resolution;
- Equipment-related artefacts are not affected by Spatial Resolution.

As a rule of thumb, for the Artefact-Spatial Resolution relation is best to keep the overall resolution higher (particularly in-plane resolution) while at the same time proactively avoiding partial volume artefact and supressing flow-related artefacts.

Figure 75: Spinal fixation using Metalwork (Cobalt). Sag STIR TSE MARS showing extensive susceptibility artefact in thoracic spine.

21 Metal Artefact Reduction Sequences (MARS)

Situations where metallic implants are present within the FOV or in the vicinity of it are always complicated. The strong Susceptibility artefact produced, by metal-based implants, can underline completely the image quality and diagnostic quality of all images.

In these cases it's fundamental to choose the protocol's sequences properly and adapt them accordingly, in hope of creating what we define as Metal Artefact Reduction Sequences. The objective of these sequences is to minimize the Susceptibility artefacts as much as possible.

The first and most important thing is to use is TSE sequences only. The rest of the existing sequences are, by one reason or other, sensitive or very sensitive to ferromagnetic and strong paramagnetic implants.

Using few sequences per scan is also essential by two reasons:

1. This is important in order to reduce SAR which, if left unchecked, will cause the implant to heat up excessively and possibly injure the patient;
2. Pretty much any change in the sequence's parameters to reduce the Susceptibility artefact will lead to a substantial increase in the sequence's TA.

21.1 Reducing Susceptibility artefacts

21.1.1 Choosing the right sequences

The first thing to do is to substitute any sequence that uses selective spectral saturation of tissues (e.g. FAT SAT, SPAIR), by another one that can give similar information. Considering that spectral tissue saturation is often difficult to acquire with decent quality using a large FOV, low B_0 strength or even near the bore's wall, it's just laughable to think that it's possible to get a decent image when there is something of significant size, in our FOV, that is effectively scrambling the proton's Larmor frequency. Even if some tissue appears saturated, there is no assurance that it's even the type of tissue that we were trying to saturate (Figure 58).

STIR and FLAIR sequences on the other hand use IR to saturate tissues and so are more resistant to Susceptibility artefacts. Since fat is the tissue that is most often saturated, STIR sequences are always a good choice for MARS protocols. This is particularly true in MSK MRI because it gives an image contrast somehow comparable to T2 with fat saturation. The artefact will still be quite significant, but tissue saturation won't be compromised too much.

As indicated earlier, always choose TSE sequences. As explained in subchapters 0 & 14.3, GE (or Hybrid) sequences are much more sensitive to Susceptibility artefacts when compared to SE sequences. It is very unlikely that acquiring images with diagnostic quality using GE sequences will be possible if the implant has significant size. Figure 76 exposes how important the type of material used is to the severity of the artefact created as well as the unsuitability of the GE sequences for imaging metal implants (although GE is great to detect the presence of metal, even tiny bits).

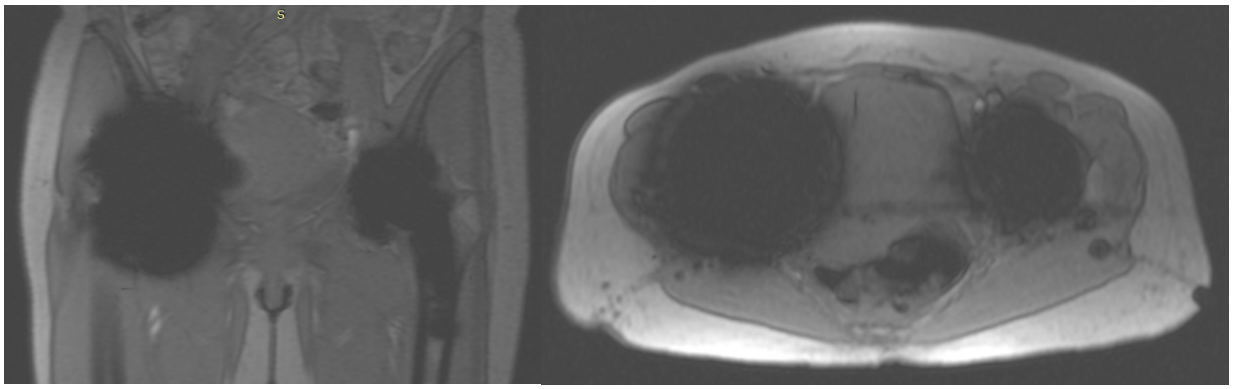


Figure 76: Coronal and Axial T1 Spoiled GE (FLASH_2D) of a human female with two hip replacements. The right hip replacement is a metal-on-metal joint replacement. The left is a ceramic-on-metal joint replacement. Right hip implant is physically smaller in size than the left hip implant, yet because of the type material is more ferro/paramagnetic the artefact created is significantly larger.

Cartesian acquisition is the best form of K-Space filling for MARS due to its low sensitivity to susceptibility, high spatial resolution, ability to generate all possible Weights, and compatibility with the most MR parameters.

Virtually all adaptations needed to transform a normal sequence into a MARS sequence will cause a substantial decrease in SNR. This inconvenient side effect must be compensated by further parameters changes. Ultimately it's the TA of the sequence that suffers the most, and so a compromise between artefact reduction, SNR and TA needs to be set.

SAR will always be a major concern in MARS because virtually all necessary parameter changes will lead to improved SAR, especially if the sequence's TA is not long. Additionally, the presence of metal implants in the vicinity of the FOV produces over-estimation errors in the pre-acquisition SAR calculation performed by the scanner and

21.1.2 Adapting parameters for MARS

The most known parameter adaption for MARS sequences is the need for very high receiving BW (> 300 Hz). A high rBW value relates to a high rate of sampling and a larger range of sampled frequencies. This allows for less chance for frequencies, with similar values (e.g. Water and Fat), to be incorrectly sampled as the same frequency and thus it reduces Chemical Shift per Pixel. Susceptibility artefacts are also reduced because the geometric and signal distortions, which are the basis of this artefact, are directly dependent on the area of "signal uncertainty" (Chemical Shift per Pixel). From this it follows that: as Chemical Shift per Pixel gets smaller, the more constricted in space the distortion will be.

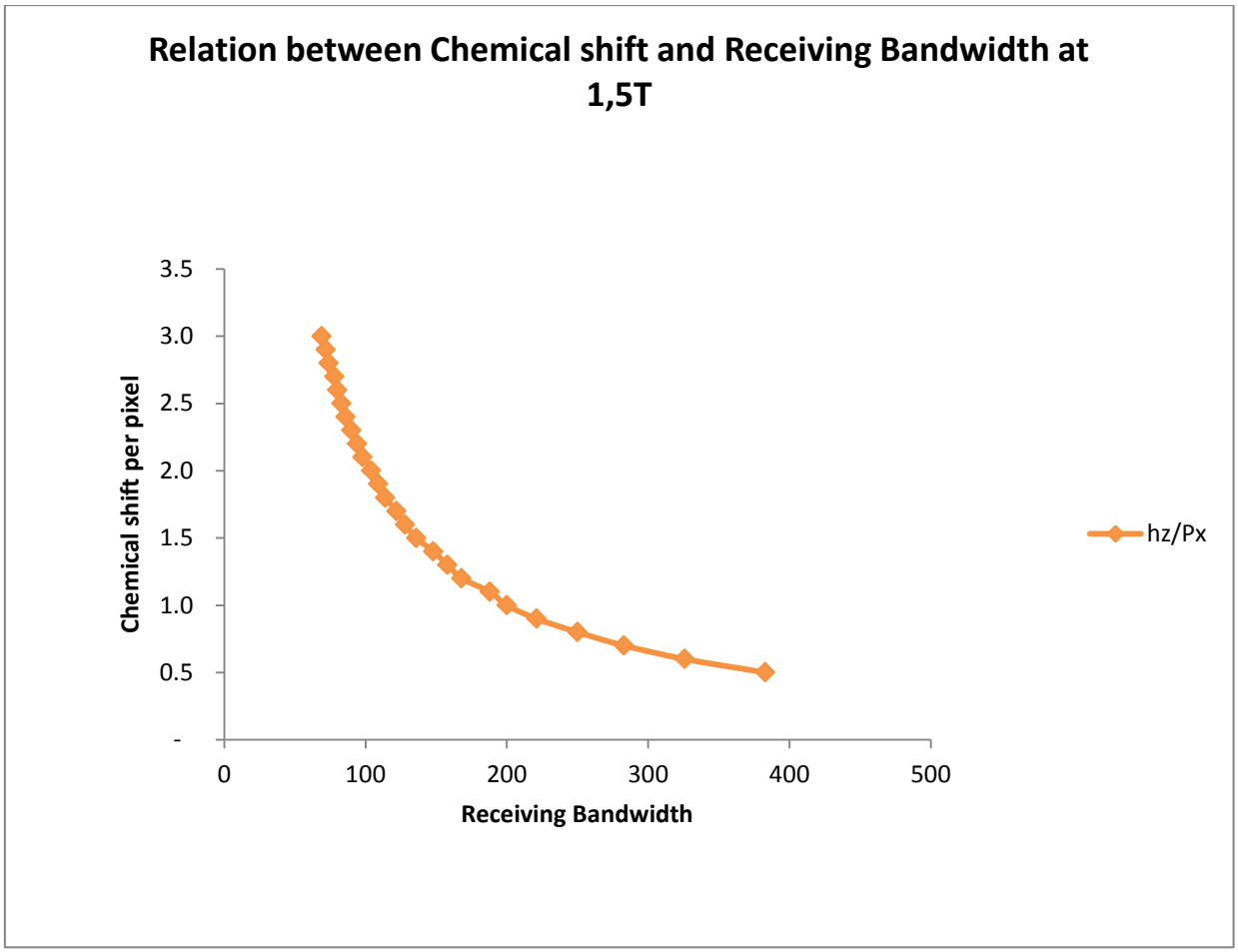


Figure 77: Above 300 Hz/Px the decrease in Chemical Shift becomes almost irrelevant. Values extracted from a 1.5T Siemens Symphony with TIM Technology.

In discussion with fellow MRI Radiographers this is often the only parameter that it's mentioned as requiring adaptation (with obvious subsequent SNR correction through additional oversampling or NEX). Changing rBW is actuality just the start of MARS.

Beginning with the receiving BW itself: setting it to above 300 Hz doesn´t have much practical meaning. Depending on the field strength of the scanner, such value for rBW can represent different values of Chemical Shift per Pixel. It is this last parameter, and not rBW, that can actually provide some degree of (non-B_0-dependent) objectivity, when reducing Susceptibility artefacts. In MARS, Chemical Shift per Pixel should be as low as reasonable allowed by SNR: ≈0.6 mm of shift per pixel (Figure 77 and Figure 78). At 1.5T, this equals to an rBW higher than 300 Hz, but at 3T the value of rBW will have to be more than the double of that value (> 620Hz per pixel for similar Chemical Shift). Although some literature [7] advises that the value of Chemical Shift per Pixel should be ≤0.5, the reality is that values below 0.7 start to require a large variation in rBW to produce a small variation in Chemical Shift per pixel (Figure 77).

Additionally SNR decreases with rBW increase in a similar mathematical fashion (Figure 78), but with higher variance for the SNR decrease.

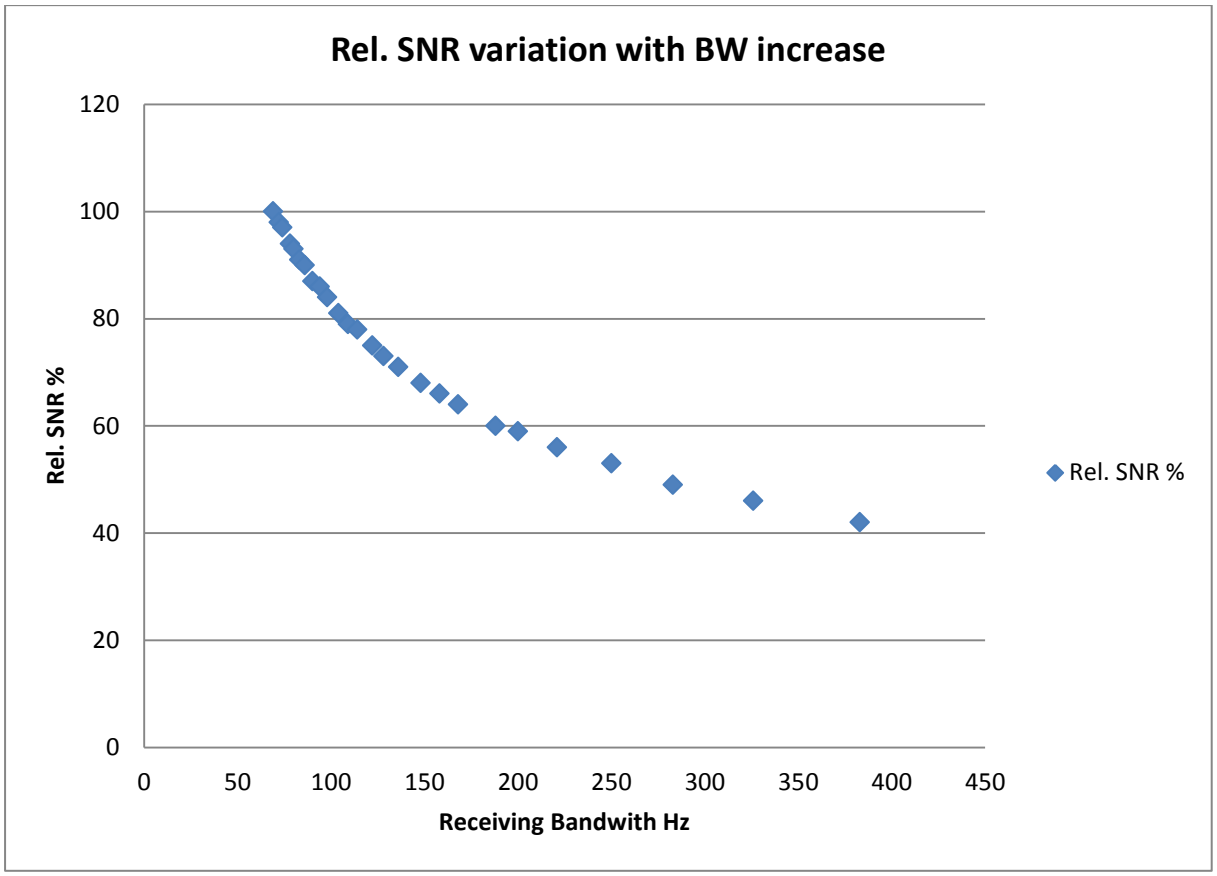


Figure 78: Relation between Relative SNR and receiving BW. The variation is quite substantial until it tends to plateau at very high bandwidth values. Keeping the rBW as low as reasonable becomes attractive in order to obtain high SNR images. Values obtained using a 1.5T Siemens Symphony with TIM Technology.

Smaller voxels represent a smaller chance for propagation of susceptibility artefacts. This implies that reducing the slice thickness and increasing in-plane resolution will diminish the artefact. Higgins, D. [7] exposed that in-plane resolution is one of the parameters that has less significance in reducing metal-related susceptibility artefacts and suggested that it should be the factor that can be neglected if the TA in MARS becomes too high. My personal experience agrees with Higgins D. [7] on the topic of high TA of MARS requiring (usually) the reduction of in-plane resolution, but for different reasons. I believe that changing the resolution (in-plane and through-plane) is the best

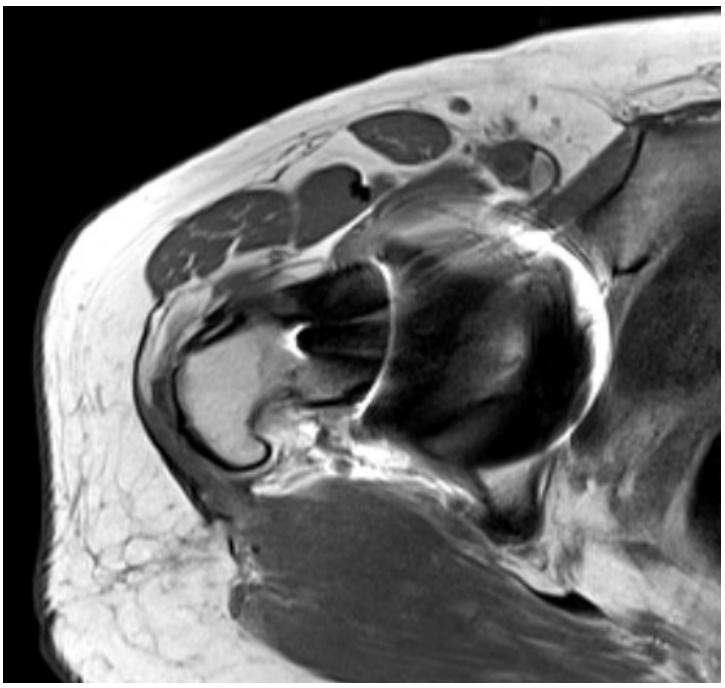


Figure 79: Axial T1 TSE of a metal-on metal hip replacement. TR=730ms; TE=7.2ms; ET=7; BW=326 Hz; Voxel Resolution =0.7*0.6*4mm; TA=3min 34s (38 slices).

approach to the problem, not because it has less relevance for artefact reduction, but because these parameters are more flexible to manipulate, than other MARS-relevant parameters. Additionally, at the cost of small/medium image resolution degradation, a substantial SNR increase and TA reduction is obtained simultaneously.

Using IPAT techniques is non-advisable because these techniques rely on using the coils' sensitivity to correctly allocate aliased signal. Strong susceptibility can compromise the signal distribution to a point where the coil´s sensitivity profile is not correct. Additionally, IPAT techniques always substantially reduce SNR, which is already very low when metal structures are present and high BWs area employed. In my experience, I have noticed that SENSE is particularly sensitive to this problem, while GRAPPA can sometimes be used with low AF. Partial Fourier is a good alternative to IPAT, when the SNR and Profile Order allow it to be used.

Avoid using RESTORE/DRIVE because, on protons located in the vicinity of metal implants, the assumption that the refocusing pulses correctly refocus MT_0 cannot be assured [7] leading to unknown contrast alterations.

TE should be as short as possible. Magnetic susceptibility accelerates the rate of T2 decay, by disturbing B_0. As a result, sampling at longer TE(s) will lead to very low SNR in the vicinity of the metal implant. Keeping TE as short as possible will reduce the effect of the accelerated T2 decay and reduce the dark area (signal void) surrounding the implant (Figure 79).

Reducing the total number of 90º RF pulses (TRs) employed and increasing the number of 180º RF pulses will also reduce areas of signal void (due to the refocusing properties of the 180º RF pulse). To achieve this, Higgins [7] suggests that ET length should be around 4*TE. This is not just impractical (since the effective ET's length is not given by all scanners and it would also lead to a long TA) but it would also limit the number of TF per TR to a short value (unless ES is extremely short). I suggest that both TF and ET should be intermediate (7 to 10 TF) or long (>12 TF) for both artefact and TA reduction. Blurring effects still need to be taken into account (sharpness is key in MARS).

It is important to bear in mind that ET increase should not be done at the expense of lengthening the ES, since that will reduce the signal intensity of the last echoes sampled and increase the artefact.

Just like TE (and for the same reasons), the ES should be as short as possible. When rBW is increased to MARS levels, the ES and minimum TE will significantly decrease because the Dwell Time is decreased. This is true for all profile orders (except Asymmetric Profile Order, because in this case ES is independent from TE). ES can be further decreased with improving other parameters (indicated below).

As exposed in subchapter 9.2.3, one important inconvenient of short ES is the occurrence of J-decoupling that is translated into images as enhanced signal intensity for fatty tissues (over water-based tissue) (Figure 80). This is turn produces obvious loss of contrast and brightness unbalance (Figure 80). This is an unwelcome side-effect of MARS sequence but one that cannot be avoided, only partially mitigated through image Normalization (Figure 80).

Using Pre-Scan Normalization/CLEAR is also not-advised by the literature [7] by the same reason as IPAT. The drawback of not using them is brightness unbalance in MRI images. This is especially true on images with large FOV or when there is a significant amount of adipose tissue. My personal experience disagrees with the literature on this point. Normalization of MARS images is important because it ameliorates the loss of contrast observed in MARS sequences, over non-MARS sequences (Figure 80), and improves the image's overall visual appeal.

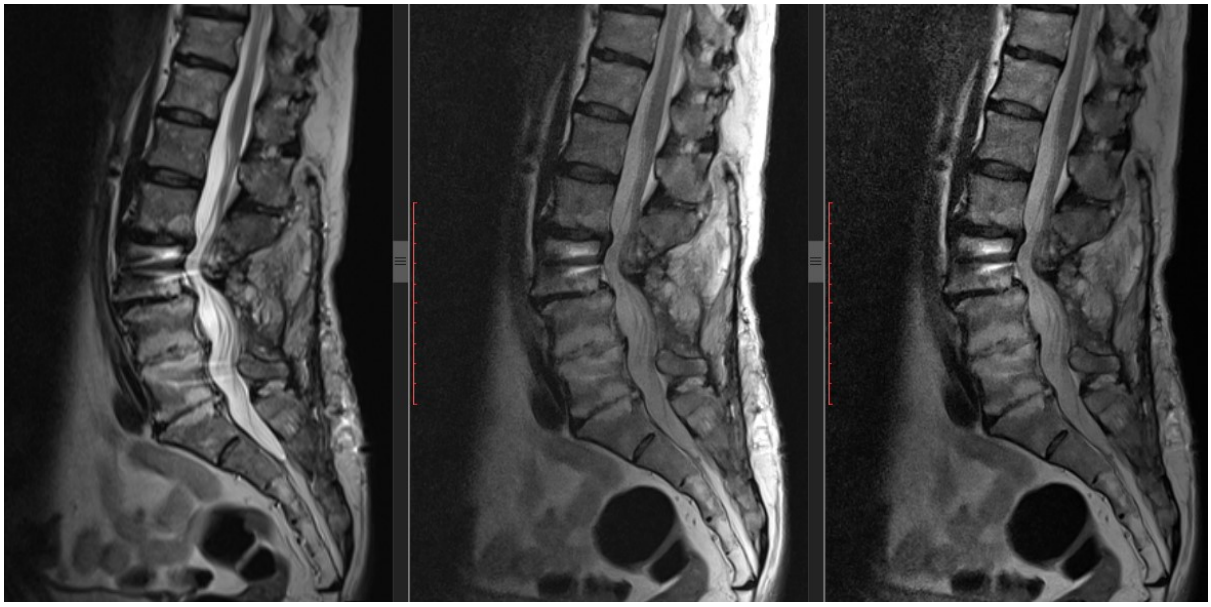


Figure 80: Sagittal T2 TSE of a human with metallic spinal fusion. Right image - Normalized Non-MARS sequence; Middle image - MARS sequence without Normalization; Left image - MARS sequence with Normalization. The loss of contrast within the CSF on the MARS sequences is evident but is less significant when Normalization is employed. Noise and the artefact intensity (not size) are also enhanced.

Improving the performance/mode the RF excitation pulse and gradients will also provide further reduction of the susceptibility artefact, but as exposed in subchapter 9.2.8.1 it will come at the cost of SNR loss and significant SAR and noise increase. ES will decrease when the performance of gradients is improved because the gradient's Rising Time decreases.

Peripheral Nerve Stimulation is more likely to occur when improved performance of gradient and RF excitation pulses are employed. Yet the likelihood of it occurring with significant severity is still low.

The idea of using Sat Bands to suppress any incorrect signal seems at first to be a simple and effective way to deal with the problem. Nothing can be farther from the truth. Sat Bands don't supress susceptibility artefacts, plus there is potential for doing more harm than good. Metal implants will distort the magnetic field, so putting a Sat Band in or in the vicinity of the metal implant may force the sat band to bend away from the implant and potentially saturate part of the area of interest. Not using Sat Bands is advisable when metal-based susceptibility artefacts are expected, plus it will allow for some reduction of the minimum TR and, by association, decrease TA.

Some manufacturers have currently on the market, some optimized sequences for scanning metal structures (e.g. WARP by Siemens). These sequences have adaptations (e.g. VAT[7], MAVRIC[8], SEMAC[9], Multiband excitation,) that go beyond the parameters manipulated by the operator and that, when combined with the adaptations already discussed, can further reduce metal-based susceptibility artefacts to very satisfactory results.

22 Image assessment

The assessment of the final images, obtained after optimization of the sequences, was performed by comparative analysis with the images routinely obtained at the NOC. All information regarding the images parameters was removed before the assessment, in order to guarantee that the assessor's choice was solely based on what they perceived on the visual information.

Seventeen cases were created for image quality evaluation, each one representing a different body structure (e.g. Lumbar Spine). Each case was composed of two groups: one contained images from the NOC's original sequences and another from the optimized sequences. Both groups represented the same clinical protocol, containing the same MR Weights.

A questionnaire (see pages 147 -155) was given to every assessor in order to standardize the image quality aspects that the assessor should focus on. The questions were equal for each presented case and covered assessment of:

- Diagnostic Quality;
- Image Contrast;
- Image Weight;
- Spatial Resolution;
- Presence of artefacts.

Images were displayed ion a DICOM Viewer (RadiAnt™ DICOM Viewer), freely available at: http://www.radiantviewer.com/download.php.

For better evaluation, the assessors were allowed to scrolls through all images/sequences, changing the images windowing, changing zoom, view multiple sequences/sequences simultaneously, etc.

Image quality evaluation was performed by 6 MSK Consultants Radiologists, 1 Radiology Fellow, 1 Radiology Registrar and 7 MRI Radiographers (2 of which were excluded. 1 was excluded for not answering the giving questions in an acceptable fashion that allowed for posterior statistical analysis. 1 was excluded for failing to assess all sequences, given for a specific body part, prior to answering the questionnaire.)

[7] VAT – Variable Angulation Tilt can reduce metal-related artefacts in in-plane direction.
[8] MAVRIC – Reduces metal-related artefacts in the though-plane direction.
[9] SEMAC – Reduces metal-related artefacts in the though-plane direction.

All opinions/comments were anonymous and fully recorded. Questionnaires filled by Radiographers were separated from the ones done by radiologists on the question of which group would the assessor prefer to report with, because none of the Radiographers were trained in MRI reporting. The Radiographers' answers to this question were recorded, but ignored for the final results regarding diagnostic quality of the images. This was done because the different skills between Radiologists and Radiographers might cause a differentiation in way the assessor analyses the images, which would lead to a distortion of the overall opinion on image quality and diagnostic quality of the images.

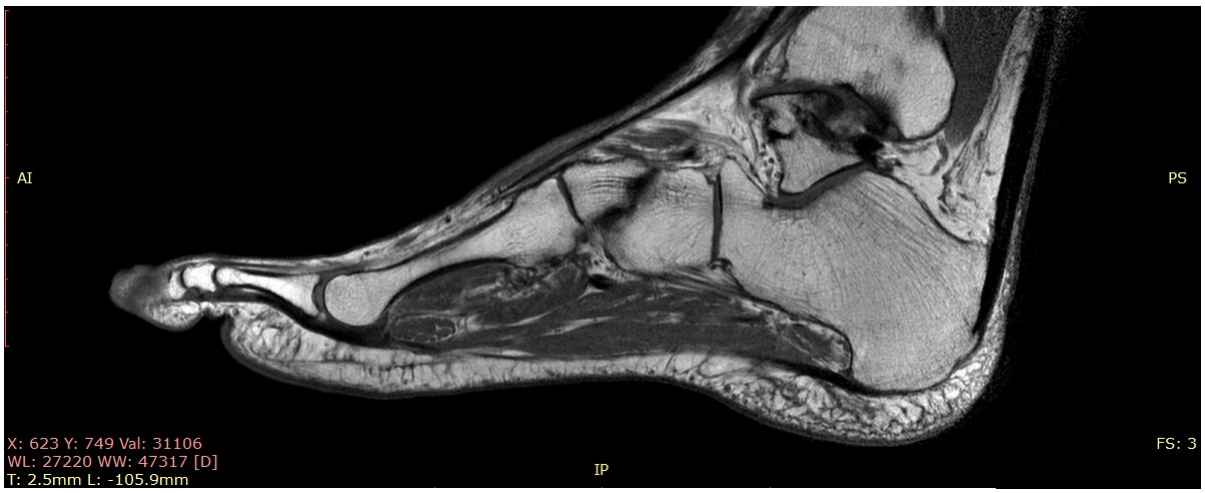


Figure 81: Sagittal T1 TSE of foot acquired with optimized sequence. Total TA: 3 min & 10 seconds. Details: Slice Thickness: 2.5 mm; Slice Gap: 0.3 mm; Acquisition Matrix: 728x726; TR: 729 ms; TE: 7.9 ms; ES: 7.9 ms; TF: 12.

Section 4: Results

23 Assessing results

After the conclusion of the optimization process, the time difference between the original protocols and the optimized protocols was compared on a case-to-case basis and overall scale, in order to assess how significant, if at all, was the time reduction.

As indicated in the previous chapter the image quality of the final images was evaluated by fellow MRI Radiographers from the NOC and the Radiologists of the NOC, by visualization on the images in standalone mode and by comparison with images from original sequences. Each opinion was registered and reviewed in order to conclude what impact the optimization process had in the overall image quality.

Statistical analysis of the results was performed using SPSS (IBM Corp. Released 2010. IBM SPSS Statistics, Version 19.0. Armonk, NY: IBM Corp) and Excel (Microsoft Office Professional Plus, version 14.0.7163.50000, Microsoft Corp).

Due to the nature of both the variables and questions posed in this project, the statistical significance of the results was calculated using the non-parametric Chi-Square test, for each of the technical IQ aspects evaluated: Diagnostic Quality, Contrast, Spatial Resolution, SNR, Dubious Image Weight and presence of Artefacts. Only P values lower than 0.05 were defined as statistically significant.

The accessors' preferred group of images (Optimized or Non-optimized) was defined by the Mode from all answers registered in the questionnaires. To further validate the results, the percentage of times that the preferred group was chosen was also calculated in order to ascertain if this preference persisted when considering only accessors with extreme levels of experience in analysing MRI images (Consultant Radiologists).

Additionally, and considering that reducing the TA of the sequences was one of the most important objectives of this project, the overall TA of every examination was compared between Optimized Sequences and Original Sequences, for several exams (see Subchapter 23.2).

23.1 Image Quality Questionnaire results

The results of the questionnaires and the comments made the accessors after completing the questionnaire exposed the following:

1. All cases were considered as having diagnostic quality by the accessors;
2. On all 17 cases presented to the accessors, group B (Optimized images) was preferred more often in 10 occasions, while group A (Non-optimized images) was only preferred on 5 occasions. Both groups were deemed as equal by the majority of accessors in only 2 cases;

3. Overall, the images produced with Optimized Sequences (Group B) where considered has having better or significantly better Diagnostic Quality (Mode = Group B) over the ones produced with the Original Sequences (Group A);
    a. This preference increased slightly when only answers from Consultant Radiologists were included in the statistical analysis;
        i. When looking at the percentage of times that Group B was chosen, it improved slightly from 55.17% (all accessors) to 57.37% (only accessors who were Consultant Radiologists).
4. Optimized Images were also assessed as having higher Spatial Resolution and slightly better Contrast overall;
    a. For Contrast this preference decreased slightly when only answers from Consultant Radiologists were included in the statistical analysis (from 56.83% to 55.93%);
    b. For Spatial Resolution this preference increased from 63.12% to 70.69% when only Consultants' answers were considered;
        i. However, the increased Spatial Resolution, of Optimized Sequences, wasn't always perceived, particularly in STIR sequences.
5. SNR was considered adequate for the vast majority of cases, but neither group of images was considered statically better ($p = 0.18$).
    a. However it was possible to stablish a Mode and it favoured Group B. This was also corroborated by the percentages calculated: 54.23% for Group B and 30.99% for Group A.
6. A good number of Artefacts was observed in both sets of images (28 artefacts found for Group B & 27 artefacts for Group A).
    a. Overall numbers of artefacts were distributed equally (no reliable Mode was obtained) but Original Images were considered by accessors as having artefacts with higher severity;
    b. Motion artefact and improper fat saturation were the most common artefacts seen in the Original images;
    c. Flow Artefacts were the most common artefact seen in Optimized Images.
7. There were a minority of cases where some images where considered as having dubious Weight: 4 instances for Optimized Images and 8 instances for Original Images. In all instances this was mostly due to presence of artefacts;
    a. Doubts regarding image Weight were raised slightly more often by MRI Radiographers.
8. TSE T1 was considered as the MR Weight where greater improvements were obtained.

The Mode and percentage of times that a specific group was chosen favoured Group B across almost all technical IQ aspects evaluated, with exception of Image Artefacts, as exposed previously.

The Chi-Squared test results obtained showed that the results attained were statistically significant ($p<0.05$) for Diagnostic Quality ($p=0.03$), Contrast ($p=0.037$) and Spatial Resolution ($p=0.023$). However there was no statistical significance for SNR ($p=0.18$) and Image Artefacts ($p=0.875$)

The following table indicates the Modes and non-parametric Chi Square test results for each variable evaluated:

Table 9: Statistical results: Mode, Chi Square Test and Percentage of times each group was chosen by the accessors

Technical IQ Aspect	Mode	Percentage of times that Optimized images were chosen	Percentage of times that Non-Optimized images were chosen	Chi-Square test asymptotic significance ($p<0.05$)
Diagnostic Quality	Group B	55.17 %	29.89 %	0.03
Image Contrast	Group B	56.83 %	33.09 %	0.037
Spatial Resolution	Group B	63.12 %	25.53 %	0.023
SNR	Group B	54.23 %	30.99 %	0.18
Artefacts	Groups were equal	54.55 %	45.45 %	0.775

23.1.1 Examples of images used in the QI questionnaire

23.1.1.1 Example 1 (Coccyx T1 Sagittal)

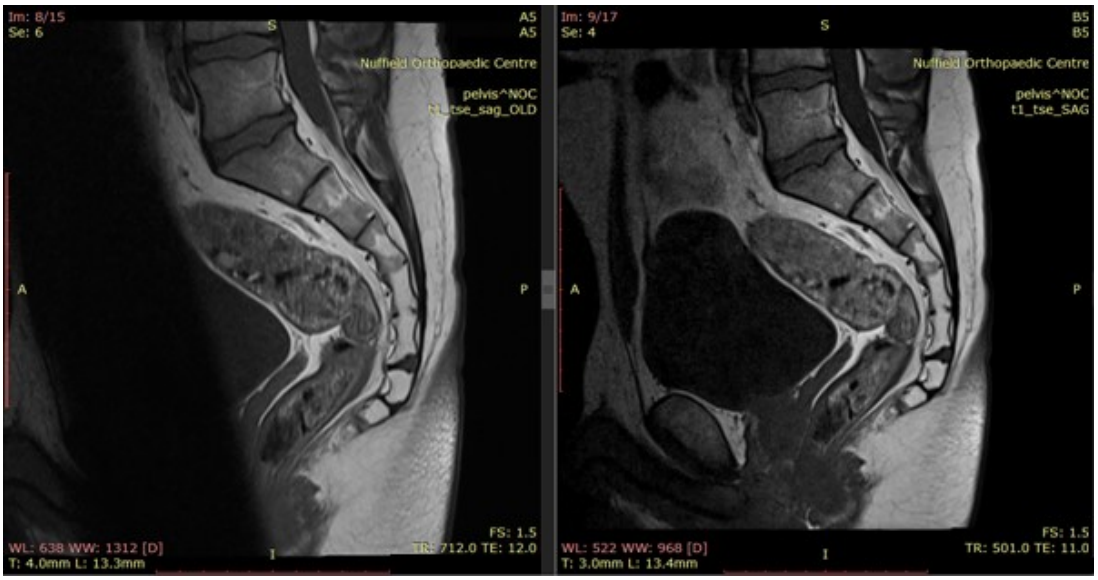


Figure 82: Both images represent a sagittal T1 TSE of the Sacrum & Coccyx acquired on a 1.5 T. Image A5 relates to an image acquired with non-optimized sequence. B5 relates to images acquired with optimized sequences. TA difference was virtually identical (around 5 seconds of difference) even though that B5 contains 2 more slices than A5. It is noticeable that the SNR is slightly lower in B5 but that the slice thickness has been reduced from 4 mm to 3mm (implying at least a SNR loss of 25% for B5). B5 also has better spatial resolution which is perceivable by the lack of smoothness in the bladder, colon and bone marrow. Contrast is also better in B5, as shown by the higher signal difference between bone marrow, intervertebral disc and CSF.

23.1.1.2 Example 2 (Sacrum T1 with Fat saturation Coronal)

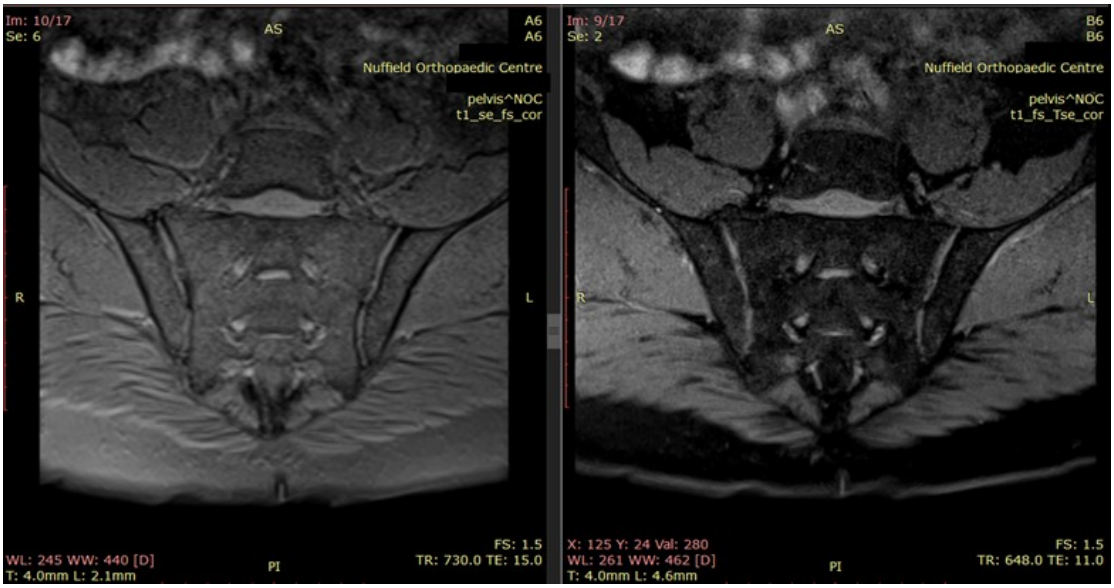


Figure 83: Coronal T1_FS TSE of SIJs. Image A6 was acquired using a non-optimized sequence and B6 was acquired using an optimized sequence. Image B6 was part of a 3 sequence protocol which compared to its non-optimized counterpart was 5 minutes and 57 seconds shorter in TA. A slight improvement (not perceivable) in the Frequency dimension of the image's matrix was introduced, along with improved SNR -TA balance and application of a full fat saturation pulse. This is why B6 has a higher noise level. This increase in noise was not considered problematic by any of the accessors. Opting for full fat saturation, instead to partial fat saturation (as seen in A6), significantly improves the image contrast and also eliminates the chemical shift artefact easily seen throughout A6, while not affecting TA significantly. However, it is important to mention that some image accessors indicated that they preferred reporting using A6 because the partial fat saturation allowed them to easily distinguish, in the SIJs, the cortical bone from bone marrow, while still being able to observe altered signal within the marrow tissue.

23.1.1.3 Example 3 (Left foot T1 Coronal)

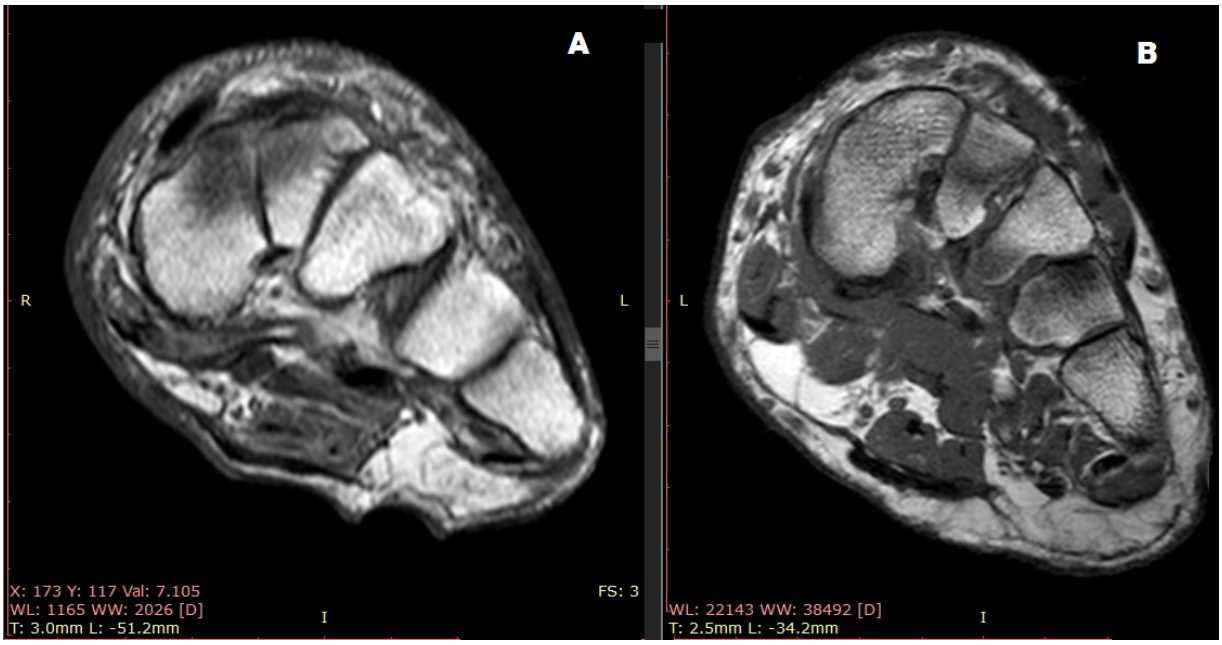


Figure 84: T1 Coronal (short axis) of the Left foot. Image A was acquired using a non-optimized sequence and image B with an optimized sequence. Even though A had a TA with over twice the length of B's TA, it produced images with substantially lower spatial resolution. In addition, the excessive amount of SNR in A produces fluctuations in signal intensity within the same tissue and throughout different regions of the FOV. Using Normalize on image A may have improved this last aspect of the image. The slice thickness is also slightly worst in A than B.

23.1.1.4 Example 4 (Left Ankle T2 Transversal)

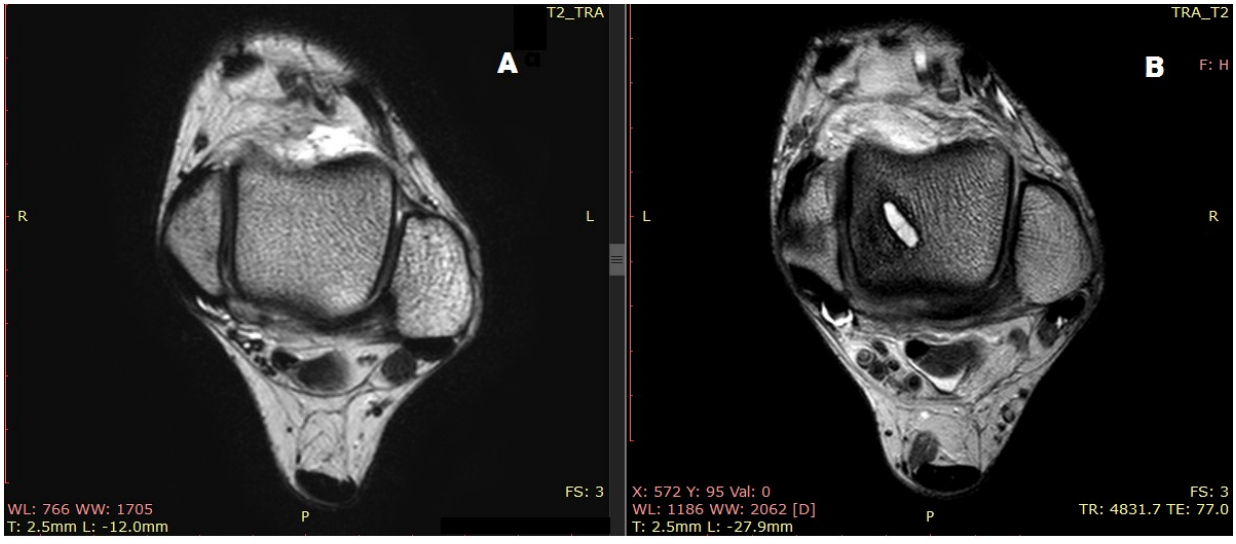


Figure 85: Right Ankle T2 Transversal. Image A (TR= 5210ms; TE=100ms) was acquired with a non-optimized sequences and image B (TR=4831; TE=77ms) with an optimized sequence. B was around 3 minutes shorter in TA than A, with the benefit of having substantially higher spatial resolution than A. Bone trabeculae is much sharped and better depicted in B than A. The substantial reduction in TA and improvement in Spatial Resolution in B was mainly attained by optimizing the echo length, employing a Drive pulse and reducing ES, TE, and TR. Using a Drive pulse allowed for a reduction on TE by 23ms without compromising the signal intensity of fluid. In turn the TE reduction produced an increase in SNR which was then trader for higher spatial resolution. Note: these images belong to two different patients of similar age and sex.

23.1.1.5 Example 5 (Lumbar spine T2 Sagittal)

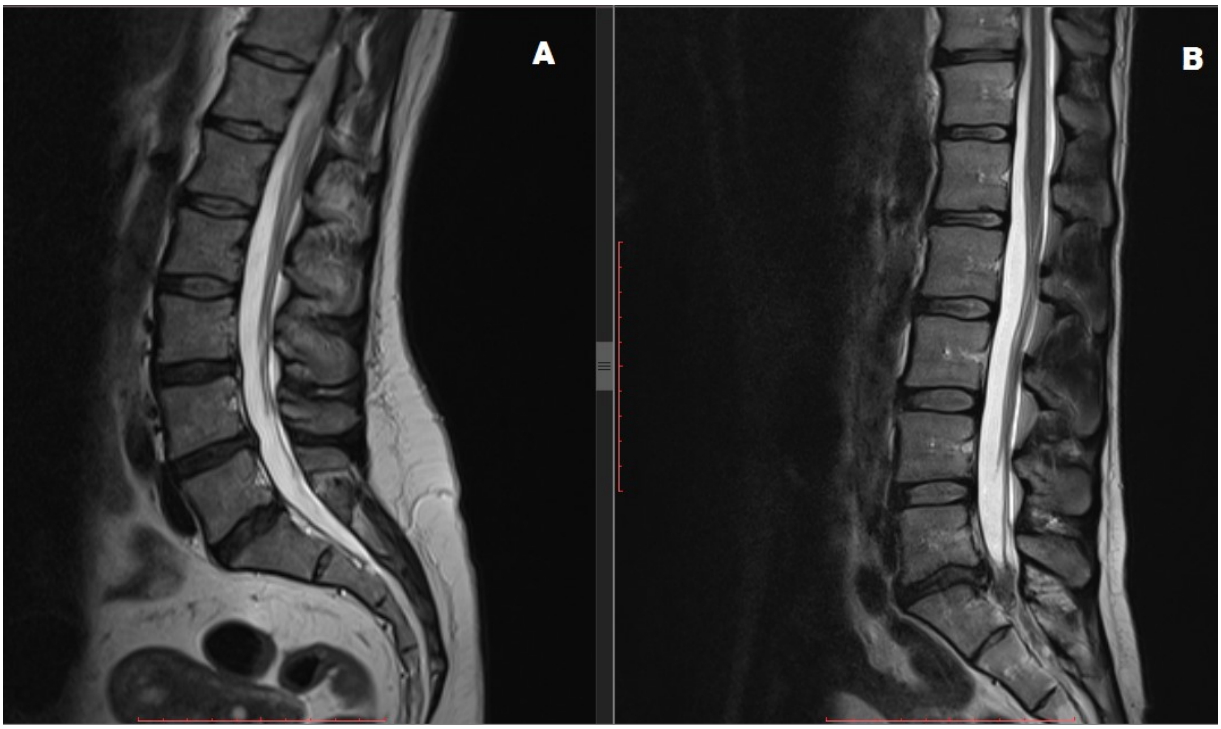


Figure 86: Sagittal T2 of Lumbar spine. Image A was acquired with a non-optimized sequence, while B was acquired with an optimized sequence. Image B shows higher sharpness and detail in relation to image A, due to using a substantial higher acquisition matrix and better SNR-TA balance. The improvement in balance on image B, in comparison to A, produced better contrast and higher sharpness throughout the entire FOV, while also allowing a reduction the sequence's TA of around 30-60 seconds. In lumbar spine scans the patient's size is a major factor to determine the distance from the receiving coils to the centre of the FOV and so large variations in intrinsic SNR should be expected. It is therefore crucial to correct these variations by improving or decreasing the percentage of oversampling for each patient. Note: these images belong to two different patients of similar age and sex.

23.2 Time reduction

After comparing the acquisition times (see Table 10) between the NOC´s Original MRI Protocols and the Optimized MRI Protocols (created in the development of this project) the following conclusions were observed:

- On average, a time reduction of 6 minutes and 48 seconds per examination was obtained;
- The protocol that showed greater reduction in acquisition time was the Arthrogram of the hip on 3T scanner, with a time reduction in the order of 17 minutes;
- The protocol that showed lower reduction in acquisition time was Coccyx fracture, with a time reduction in the order of 30 seconds;
- On average, time reductions were higher in longer/more complex examinations (e.g. Hip arthrogram);
- The sequence that showed greater TA decrease was TSE T1.

Table 10: Time length of optimised MRI examinations

Body part	Protocol	Total original acquisition time	Total optimized acquisition time	Time difference	Absolute time difference
C spine	routine/radiculopathy	00:11:24	00:07:40	00:03:44	00:03:44
	MARS	00:23:12	00:15:24	00:07:48	00:07:48
	limited	00:08:13	00:05:08	00:03:05	00:03:05
	Limited +axials	00:13:25	00:07:58	00:05:27	00:05:27
	avulsed roots	*	00:12:17	#################	00:12:17
	trauma	00:13:36	00:07:58	00:05:38	00:05:38
	tumour	00:18:02	00:12:55	00:05:07	00:05:07
	Paediatric Torticollis/Rotary subluxation	*	00:11:33	#################	00:11:33
C/T spine	routine/radiculopathy	00:22:05	00:12:06	00:09:59	00:09:59
T/L spine	limited	00:07:56	00:05:34	00:02:22	00:02:22
	Limited +axials/trauma	00:22:38	00:13:56	00:08:42	00:08:42
	MARS	00:17:41	00:11:04	00:06:37	00:06:37
	vertebroplasty pre assessment	00:17:11	00:08:05	00:09:06	00:09:06
	routine/radiculopathy/ CES 3T	00:16:19	00:10:00	00:06:19	00:06:19
T spine	routine/radiculopathy/ CES 1.5T	00:12:47	00:08:17	00:04:30	00:04:30
	MARS	00:21:42	00:14:09	00:07:33	00:07:33
L spine	Limited	00:07:35	00:04:48	00:02:47	00:02:47
	Limited +axials/trauma	00:15:07	00:08:37	00:06:30	00:06:30
	vertebroplasty pre assessment	00:17:11	00:06:47	00:10:24	00:10:24
	Tumour	00:23:16	00:13:42	00:09:34	00:09:34
	Pars Defect	00:11:32	00:05:43	00:05:49	00:05:49
Whole spine	routine/radiculopathy	00:31:15	00:20:22	00:10:53	00:10:53
	Limited	00:15:07	00:10:06	00:05:01	00:05:01
	Limited +axials/trauma	00:34:21	00:21:43	00:12:38	00:12:38
	Scoliosis	00:18:27	00:17:18	00:01:09	00:01:09
SIJs[10]	sacroiliitis/routine	00:13:11	00:07:14	00:05:57	00:05:57
	Fracture	00:07:59	00:04:18	00:03:41	00:03:41
Coccyx	coccygeal pain	00:04:57	00:04:46	00:00:11	00:00:11
Thoracic outlet	Routine	00:14:25	00:10:51	00:03:34	00:03:34
brachial plexus	Routine	00:22:40	00:15:31	00:07:09	00:07:09
Rib/Intercostal muscle injury	Side Strain	00:26:02	*	00:26:02	00:26:02

[10] SIJs – Sacroiliac Joints

TMJ[11]	routine 1.5T	*	00:20:05	#################	00:20:05
	routine 3T	*	00:15:38	#################	00:15:38
Knee	Internal Derangement >35 years old 3T	00:17:14	00:11:59	00:05:15	00:05:15
	Internal Derangement >35 years old 1.5T	00:17:28	00:11:50	00:05:38	00:05:38
	Internal Derangement <35 years old 3T	00:17:03	00:12:34	00:04:29	00:04:29
	Internal Derangement <35 years old 1.5T	00:16:52	00:12:33	00:04:19	00:04:19
	Internal Derangement >35 years old 3T for possible knee replacement	00:25:27	00:15:37	00:09:50	00:09:50
	Internal Derangement large knee	00:23:56	00:23:56	00:00:00	00:00:00
	MARS 3T	00:24:19	00:13:40	00:10:39	00:10:39
	MARS 1.5T	00:21:46	00:12:22	00:09:24	00:09:24
	Tumour 3T	00:25:23	00:18:05	00:07:18	00:07:18
	Tumour 1.5T	00:22:05	00:16:01	00:06:04	00:06:04
	Previous Meniscal surgery	*	00:15:36	#################	00:15:36
Hip	routine 3T	00:21:41	00:11:58	00:09:43	00:09:43
	routine 1.5T	00:17:10	00:10:34	00:06:36	00:06:36
	Tumour 1.5T	00:21:40	00:13:34	00:08:06	00:08:06
	MARS 1.5T	00:22:26	00:12:24	00:10:02	00:10:02
	Slipped Upper Femoral Epiphysis	00:15:06	00:17:10	#################	00:02:04
	Arthrograms 1.5T	00:33:44	00:33:44	00:00:00	00:00:00
	Arthrograms 3T	00:41:08	00:23:50	00:17:18	00:17:18
	DDH Post closed reduction	00:04:06	00:03:27	00:00:39	00:00:39
	Pseudo tumour	00:22:26	00:12:24	00:10:02	00:10:02
	athletic groin pain	00:29:02	00:26:30	00:02:32	00:02:32
	non-gadolinium Arthrograms	00:47:01	00:38:37	00:08:24	00:08:24
Calf	Calf/Tibiae stress fractures	00:15:23	00:14:18	00:01:05	00:01:05
tumour	soft tissue tumour	00:20:35	00:15:51	00:04:44	00:04:44
leg/thigh	bone tumour	00:20:35	00:15:51	00:04:44	00:04:44
	tumour follow up	00:24:37	00:17:39	00:06:58	00:06:58
infection	Myositis	00:16:57	00:12:37	00:04:20	00:04:20
	Osteomyelitis	00:16:57	00:12:37	00:04:20	00:04:20
ankle	routine 3T	00:33:26	00:19:25	00:14:01	00:14:01
	MARS	00:36:13	00:22:34	00:13:39	00:13:39
	Tumour	00:32:30	00:22:08	00:10:22	00:10:22

[11] TMJs - Temporomandibular Joints

	Coalition	00:31:14	00:20:31	00:10:43	00:10:43
	Infection	00:20:39	00:12:03	00:08:36	00:08:36
Foot	routine 3T	00:33:10	00:25:58	00:07:12	00:07:12
	routine 1.5 T	00:24:55	00:17:20	00:07:35	00:07:35
	soft tissue infection	00:27:28	00:13:50	00:13:38	00:13:38
	Morton's neuroma	00:15:10	00:15:10	00:00:00	00:00:00
	stress fracture	00:18:57	00:13:33	00:05:24	00:05:24
shoulder	routine 3T	00:20:47	00:15:31	00:05:16	00:05:16
	routine 1.5T	00:15:13	00:11:24	00:03:49	00:03:49
	Arthrograms	00:38:57	00:22:23	00:16:34	00:16:34
	Tumour	00:29:02	00:16:52	00:12:10	00:12:10
Wrist	Routine	00:19:25	00:17:05	00:02:20	00:02:20
	Arthrograms	00:27:39	00:18:06	00:09:33	00:09:33
	Tumour	00:25:30	00:17:54	00:07:36	00:07:36
	MARS	00:22:37	*	#################	00:22:37
	scaphoid fracture/trauma	00:19:21	00:10:02	00:09:19	00:09:19
	scaphoid AVN	00:19:21	00:10:02	00:09:19	00:09:19
hand/fingers	routine hand	00:28:52	00:24:58	00:03:54	00:03:54
	Thumb	00:27:00	00:23:14	00:03:46	00:03:46
	Tumour	00:30:50	00:24:49	00:06:01	00:06:01
	routine finger	00:28:10	00:22:05	00:06:05	00:06:05
	Synovitis	00:22:00	00:22:00	00:00:00	00:00:00
elbow	routine 3T	00:32:27	00:17:34	00:14:53	00:14:53
	routine 1.5T	00:27:47	00:18:23	00:09:24	00:09:24
	Arthrograms	*	00:21:39	#################	00:21:39
	Tumour	00:25:04	00:20:07	00:04:57	00:04:57
sterno-clavicular joints	Routine	00:19:34	00:14:00	00:05:34	00:05:34
	tumour (3 planes)	00:29:41	00:20:03	00:09:38	00:09:38
Pelvis	routine 3T	00:17:36	00:16:57	00:00:39	00:00:39
	routine 1,5T	00:22:23	00:13:15	00:09:08	00:09:08
	tumour 1,5T	00:25:08	00:16:28	00:08:40	00:08:40
Average time reduction with Optimized Sequences:					**00:06:48**

* Protocol non-existent

Section 5: Discussion & Conclusion

24 Discussion

The range of MRI examinations within the speciality of MSK is vast and varied. This renders a detailed assessment of every body part impossible for a project that is not joint specific. Such analysis would require a medium or large number of samples for each body part assessed, for both the NOC's Original Sequences and for Optimized Sequences. For this reason an array of multiple types of MSK examinations were chosen for assessment, instead of a large number of exams from a single type of MSK examination. It is true that the latter would provide better insight into the specific type of examination chosen and it also would allow for better assessment of the protocol's robustness to different patients/pathologies/positioning variations. However, only the first approach could provide an adequate overview of the overall image quality produced by the scanner and expose how effective sequence optimization can be in a width variety of cases and scanning conditions.

The number of image assessors employed in this project not only vastly exceeds the average number of assessors used in other published clinical and technical studies, that include visual analysis of images, but it also managed to find assessors of different backgrounds and with diverse degrees of experience in MRI. This came about as a result of the author's initial intent of observing how MRI images are perceived and interpreted by assessors with different backgrounds and experience. It was also done to ensure that the sample of images evaluated was sufficiently large without requiring each assessor to spend an extensive period of time answering the questionnaire.

It was noticed that Consultant Radiologists were quicker at assessing the images and had less doubts regarding image Weight, Contrast and Diagnostic Quality. This was expected considering that they possess the greatest experience in assessment of MRI images among all groups of assessors. However, it was curiously noticed that Consultant Radiologists were worse than MRI Radiographers at assessing/deciding which images had better SNR and at discovering Artefacts within the images. They also expressed some difficulty in discerning which group of images had higher Spatial Resolution in cases where this parameter's change wasn't very obvious between the two groups of images.

The answers from Radiologist Fellows and Registrars weren't compared with the rest of assessors due to the low number of individuals with these backgrounds.

The fact that T1 was the Weight where greater image quality improvement was achieved with sequence optimization is of high significance because of the importance that T1 Weight has in MSK MRI. Associated with this, T1 was also the sequence where greater TA reduction was obtained, with a difference in some cases of more than 2 minutes. This was possible by exploiting the excessive SNR found in the original T1 sequences and also by swapping T1_SE T1 for T1_TSE, whenever the first one was used. Higher Spatial Resolution was the main factor that contributed to improvement of T1 image quality.

Improving the Spatial Resolution of STIR sequences, with medium/large FOVs, was considered of poor relevance by most assessors for improving its diagnostic quality. Improvement of STIR sequences should then be more focused in guaranteeing high contrast, reducing TA and reducing/eliminating artefacts. The more prevalent artefacts in STIR were in-plane flow artefact and susceptibility-related artefacts

Reduction of artefact severity one of the main factor that contributed to the improvement of both image and diagnostic quality in the images produced from Optimized Sequences. On the Original Sequences improper fat saturation and motion artefact were the most common artefacts seen. By contrast, on Optimized Sequences the single most common artefact seen was in-plane artefact of mild or low severity. Improper fat saturation was also present in some cases but with much lower severity.

The assessor's overall preference for images produced with Optimized Sequences in relation to Diagnostic Quality and Contrast is proof that the measured reduction in TA wasn't attained at the cost of image quality or distortion of the clinical information contained within MRI images.

From the sacroiliac joints examination included in the questionnaire an interesting question popped up: "Is full fat saturation preferable to partial fat saturation?" Answers from Consultant Radiologists were mixed, with the majority preferring partial fat saturation for situations were assessing the thickness of both cartilage and cortical bone is fundamental. Full fat saturation was preferred when looking for diffuse signal changes within marrow tissue due to the higher contrast. The minority preferred full fat saturation in all situations and justified the desire for it on the sacroiliac joints with the same reasons given by the majority. This apparent contradiction originates from how different consultants perceive and assess the images. While some are more focused in looking for thickening of the cartilage, others are more interested in locating areas were cortical bone is becoming thinner. Unfortunately, this project did not allow for further and more adequate investigation of this question. However, given the importance and prevalence in the use of fat saturation in MSK MRI, this question deserves further dedicated research.

25 Conclusion

This project shows that it's definitely possible to improve image quality and reduce the length of MRI examinations, simultaneously. There are obviously limits and making compromises is never simple but, if done correctly, it can add an important edge in the provision of healthcare MRI services.

The increase in image quality was obtained through improvement of multiple aspects. Among them the main factors were: higher spatial resolution, better contrast and increased image sharpness.

As observed by the Radiologists' comments, registered during image assessment, the image quality improvement translates into improved diagnostic quality and increased Radiologist confidence while reporting.

The 6 minutes and 48 seconds of overall time reduction per examination attained is an important point to bear in mind, since it can be translated into an MRI scanner output increase of 1 to 4 patients per day.

Optimizing MRI sequences is a complex and lengthy process that requires few resources, but a lot of dedication, time and focus in order to be done.

The added value that the optimization process can bring is dependent on the desire/objective of whoever performs it, since it can be done simply to improve one specific aspect of MRI scanning (e.g. Image's Contrast) or multiple aspects. It can then be said that MRI image optimization is a flexible process that can be tailored, to a certain extent, to the user's needs. This characteristic increases the range of possible scenarios where MRI image optimization can be employed.

It was noticed during this work, that both SAR and the Acoustic Noise produced by Optimized Sequences were significantly higher. Nevertheless, because of the safety restrictions all MRI scanners possess, the safety limits of both SAR and Acoustic Noise were always respected (even in MARS sequences). No further assessment of these factors was done because they were not part of this project's objectives and the lack of adequate measuring equipment.

Considering the results of the questionnaires, the NOC's MRI protocols have been updated. There is no change on the type of sequences or image Weights that compose the protocols, but now the MRI protocols are composed from Optimized Sequences. This has led to reduced scan times and fewer complaints from Radiologists regarding the diagnostic quality of the images. This is especially true in cases where metal implants are present in the images.

Section 6: References, Appendixes and other Accessory Information

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Appendix 1. Table of parameters relations/trade-offs

Table 11: Parameters Relations/Trade-offs

Parameter	Time	Spatial Resolution	SNR	Artefacts	Contrast
TR	+	=	=	=	Less T1 weight
TE	=	=	=	=/+/-	More T2 weight
TI	+	=	=	=	Partial or total saturated of a specific tissue
Concatenations/Packages	++	=	=	-	Only if TR in changed
SE flip angle	=	=	-	=	Less T1 weight at higher FAng
GE flip angle	=	=	-/+	=	More T1 weight at FAng over 50º Less T1 weight at FAng under 40º
NEX	++	=	++/+	-	Increase in overall CNR
Voxel size	-	-	+	+/-	Increase in overall CNR
Read FOV	=	--	++	=/-	Increase in overall CNR
Phase FOV	+	=	+	=/-	Increase in overall CNR
Slice Thickness	=	--	++	=	Increase in overall CNR
Number of slices (2D)	++	=	=	=	=
Slice gap	=	=	=	=/-	=
Read Matrix	=	+	--	=	Decrease in overall CNR
Phase Matrix	+	=/+	-	=	Decrease in overall CNR
Sat Bands	+	=	=	-	Increase in overall CNR by artefact reduction
Half/Partial Fourier	--	=	--	+	Decrease in overall CNR
Chemical Shift	=	=	+	++	Decrease in overall CNR
Bandwidth (receiving)	-	=/+	-	-	Increase in overall CNR by artefact reduction
Bandwidth (transmitting)	-	=	=	-	=
Gradient mode	-	=	=/-	-	=

Parameter	Time	Spatial Resolution	SNR	Artefacts	Contrast
Turbo factor/Echo train	-	=/-	=	=/+	Blurring effect may appear, reducing CNR. Fat signal becomes hyperintense.
Echo spacing	+	=	=	+	=
3D slabs	+	=	+	=	=
Flow/motion correction	=/+	=	--	-	=
Partial echo	=	=	+/-	+/-	=
Fat Saturation	+/++	=	--	+/-	Fat tissue partially of totally saturated
IPAT	--	=	--	+/=	=
Interpolation	=	++	=	=/+	=
Oversampling	+	=	+	-	=
Restore magnetization/Drive	+	=	+	=	Accelerated relaxation of M_L. Increase in intensity value of fluids for short TRs
2D/3D Distortion correction	=	=	=	-	=
Filters	=	-	=	=	Increase in overall CNR by blurring effect or contrast enhancement
Triggering/PACE	+/=	=	=	--	Improved contrast by motion artefact reduction
Normalize	=	=	=	-	Very slight reduction by Image brightness homogenization throughout entire FOV
legend	+(increase) - (decrease)	=(unchanged) / (or)			

Appendix 2. Table of sequence acronyms for several manufacturers

Table 12: Sequence Acronyms for several different Manufacturers. Adapted from [13]

Type of sequence	Philips	Siemens	GE	Hitachi	Toshiba
Spin Echo (SE)	SE	SE	SE	SE	SE
Multi echo SE	Multi SE	Multi echo	SE	SE	Multi écho
		MSHECO			
		MS			
Fast SE	Turbo SE	Turbo SE	Fast SE	Fast SE	Fast SE
Ultra fast SE/ Single Shot SE	SSH-TSE	SSTSE	SS-FSE	FSE - ADA	(Super)FASE
	UFSE	HASTE			DIET
IR	IR	IR/IRM	IR	IR	IR
Fast SE IR	IR TSE	TurboIR/TIRM	FSE-IR	FIR	Fast IR
STIR	STIR	STIR	STIR	STIR	STIR
Fast STIR	STIR TSE	Turbo STIR	Fast STIR	Fast STIR	Fast STIR
FLAIR	FLAIR	FLAIR	FLAIR	FLAIR	FLAIR
FAST FLAIR	FLAIR TSE	Turbo FLAIR	Fast FLAIR	Fast FLAIR	Fast FLAIR
Gradient echo (GE)	FFE	GRE	GRE	GE	FE
Spoiled GE	T1-FFE	FLASH	SPGR	RSSG	RF-spoiled FE
			MPSPGR		
Ultra fast GE	T1-TFE		FGRE	SARGE	
	T2-TFE	TurboFLASH	Fast SPGR		Fast FE
3D Ultra fast GE	THRIVE	VIBE[12]	FMPSPGR		RADIANCE
			VIBRANT		QUICK 3D
			FAME		
			LAVA		
Ultrafast GE with magnetization preparation	IR-TFE	T1/T2-TurboFLASH	IR-FSPGR DE-FSPGR		Fast FE
Steady state GE	FFE	FISP	MPGR, GRE	TRSG	FE
Contrast enhanced steady state GE	T2-FFE T2	PSIF	SSFP		FE
Balanced GE	Balanced	True FISP	FIESTA	BASG	True SSFP
	FFE				
SE-Echo planar	SE-EPI	EPI SE	SE EPI	SE EPI	SE EPI
GE-Echo planar	FFE-EPI	EPI Perf.	GRE EPI	SG-EPI	FE-EPI
	TFE-EPI	EPIFI			
Hybrid echo	GRASE	TGSE			Hybrid EPI

[12] VIBE - Volume Interpolated Breath Hold Examination

Appendix 3. Soft Tissue Phantom Proposal - NOC

The following essay consists of a proposal with the objective of obtaining approval to scan non-human non-living soft tissue (ex: lamb leg or pig leg), using the NOC's MRI scanners (1.5T Siemens Symphony and 3T Philips Achieva XL).

The objective of the practical work developed, through the scanning of the object mentioned above (hereafter referred as: soft tissue phantom), is the development of optimized MRI sequences and optimized MRI protocols, for all clinical MRI examinations performed at the NOC.

The results of the work developed will be used to improve the image quality of the NOC's clinical MRI examinations and improve the scanner's throughput. The results will also be used as a basis to generate a Master's Dissertation on the subject of: "Optimization of MRI sequences in a MSK clinical setting", whose author is Cláudio Pereira.

Using a soft tissue phantom for this kind of practical work is essential, as it provides excellent similarity to scanning human patients while avoiding using human test subjects. Because sequence optimization is a complex and slow process than needs to rely on a trial-and-error methodology, using human test subjects is far more a complex management/safety/ethical affair than using a soft tissue phantom. Additionally a soft tissue phantom allows for a better control of the factors at play, removing variability like motion or flow related artefacts.

To avoid any disturbance of the regular clinical and research schedules of the MRI scanners (both NHS and private work), the work shall be developed out-of hours, when the scanners aren't in use.

Unfortunately soft tissues phantoms present themselves with the problem of being potentially harmful for a clinical environment. Due to the risk of cross infection and environment/equipment contamination, measures must be put in place to prevent potential risks.

The following list comprises of the measures to be employed to prevent any infection/contamination related hazards, in case of successful approval of this proposal:

- The soft tissue phantom will be kept isolated, at all times, from the surrounding environment by covering it in a 3 layer of hermetically sealed thick plastic bags;
- When in use, the soft tissue phantom will be nested in an absorbent base to prevent any spread of any leakage that may occur;
- When not in use, the soft tissue phantom will be kept isolated and secure in a shock resistant hard plastic case, properly labelled as: "Soft Tissue MRI Phantom. Please do not open case or move from storage location";
- To reduce the deterioration rate of the soft tissue phantom, that would increase the risk of fungi development and increase the risk of airborne dispersion of potential harmful pathogens, the encased soft tissue phantom will be kept in a low temperature climate controlled environment (Open Scanner's fridge);

- Each Soft Tissue Phantom will be stored in a controlled environment for a maximum period of 1 week. If a need of a phantom arises beyond the time length, a new soft tissue phantom will be used following the same protection measures;
- After each time the phantom will be used, the entire scan room will be thoroughly cleaned using antibacterial and anti-sporicidal disinfectant products (that are compatible with the MRI scanners);
- The work developed will be performed after hours to reduce the contact between the soft tissue phantom and the Trust's staff, patients and visiting personnel.

Regards
Cláudio Pereira

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Oxford
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Appendix 4. Standard Operating Procedure – Scanning non-living non-human tissue on *a Magnetic Resonance Imaging (MRI) Scanner*

Institution:	Oxford University Hospitals NHS Trust (OUH)	Contact:	
Site:	Nuffield Orthopaedic Centre (NOC) – Radiology	Author:	Cláudio Pereira
Person Responsible:	MRI and Research Superintendent	Date of Approval:	29th October 2014

STANDARD OPERATIONAL PROCEDURE (SOP) OUTLINE:

This SOP was written due to the need to establish guidelines that ensure safe work practice infection control wise when scanning non-living non-human biologic tissues. This need arose in response to the proposal by Cláudio Pereira, to the NOC's Radiology Department, to develop a practical study using the NOC's MRI scanners. This study is a fundamental component of the Master Dissertation currently being written by Cláudio Pereira, on the subject of: "Optimization of MRI sequences in a MSK clinical setting". It requires the scanning of biological tissue for assessment of image quality, diagnostic quality and others technical parameters.

TIME FRAME:

It is expected that the time frame for scanning won't extend beyond 2 months.

DESCRIPTION:

This document provides infection control guidelines to be followed when scanning non-living non-human tissue/objects for non-clinical purposes, in all of the NOC's MRI scanners.

OBJECTIVES:

1. To guarantee that no risk of biological contamination to staff, patients, equipment and people visiting the NOC arises from any work developed;

2. Be in accordance with all of the OUH's Infection Control Policies and Regulations.

INFECTION/CONTAMINATION HAZARDS:

Non-living non-human biological tissues (BT) present themselves has excellent vessels to harbour and grow bacterial and viral pathogens. Some of these pathogens are potential harmful to humans, particularly those with weak or suppressed immune systems (ex: children, elderly, sick individuals, etc.). Because proper sterilization of these tissues is not possible, it becomes necessary to prevent/limit the exposure of individuals (especially those with weak or supressed immune systems) to such tissues. This applies to direct exposure and to indirect exposure, by contamination of surfaces or other individuals.

To reduce any contact with patients or staff and not to interfere with daily clinical work, Scanning of BT can only be performed during out-of-hours periods.

Measures relating to the use acquisition, handling, cleaning, storage and disposal of such tissues need be defined.

ACQUISITION OF BIOLOGIC TISSUES MUST FOLLOW WHAT IS INDICATED BELOW:

- It is the responsibility of the person in charge of the study to acquire the BT;
- BT must follow the same criteria as those applied to any commercial food/drink intended for human consumption;
    - For the study referred, the BT acquired will be animal meat/pieces purchased in a commercial food store to ensure that it complies with regulation regarding foods for human consumption.

ISOLATION/PROTECTION OF BIOLOGIC TISSUES MUST FOLLOW WHAT IS INDICATED BELOW:

- Before being used in the study, BT must be properly isolated from the outside environment;
    - For the study referred, BT will be acquired already completely surrounded with a resistant leak proof plastic;
    - Additionally, the BT will be kept inside 2 other layers of airtight resistant plastic bags and 1 airtight hard plastic container;
- The isolation method must be resistant to physical shock/manipulation and leak proof;
- Isolation materials must not deteriorate with use.
- Isolation of BT must be ensured at all times;
- If any defect in the isolation materials is found, they must be replaced immediately.

HANDLING OF BIOLOGIC TISSUES MUST FOLLOW WHAT IS INDICATED BELOW:

- Latex gloves must be worn at all times when handling BT;
- All surfaces where BT is placed must be covered with disposable absorbent paper;
- All contact with BT must be non-direct;
    - Non-direct contact refers to any type of contact where there is at least one resistant protective barrier that separates the BT and the handler.
- BT must only be handled for the purposes of the study and nothing else;

STORAGE OF BIOLOGIC TISSUES MUST FOLLOW WHAT IS INDICATED BELOW:

- BT must be kept safely stored in a low temperature climate controlled closed environment (ex: fridge) at all times whenever it is not being used at the time, for the purposes of scanning it;
    - For the study referred, the fridge located in the Open Scanner will be the designated place of storage since isn't currently in use and is located in a safe area, only accessible to a very limited number of staff.
- BT must not be stored alongside food, drink, medication or other objects that can themselves became easily contaminated with pathogens;
    - The fridge mentioned in the line above is currently empty and not in use.
- BT can only be storage and used while it shows no visual/olfactory signs of degradation/decomposition/bacterial, fungal or viral contamination;

- For the study referred, BT will be changed/replaced every 5 days routinely, or sooner if contamination is detected.

CLEANING AND DISINFECTION OF EQUIPMENT AND OTHER SURFACES:

- All equipment used (scanner, table, coils, pads, etc.) must be thoroughly cleaned and disinfected at the end of every scanning session, with appropriate cleaning tools;
    - For the study referred, the cleaning tools employed will be disposable disinfectant wipes (Clinell, etc.), currently used by the NOC's Radiology Department for day-to-day use.
- Any fluid leaks from the BT will be cleaned by first absorb the excess fluid in absorbent paper followed by thorough cleaning and disinfection of the surface or object that came in contact with the fluid.

DISPOSAL OF BIOLOGIC TISSUE MUST FOLLOW WHAT IS INDICATED BELOW:

- Proper disposal of BT must be ensured by the person in charge of the study;
- The OUH has no responsibility regarding the proper disposal of BT;
- Disposal of BT must not infringe the laws and regulations of Public Health regarding this topic.

Appendix 5. Template of the Decontamination log used

Oxford University Hospitals NHS

NHS Foundation Trust

NOC – MRI Decontamination Diary (version 1.0 Jan.2015)

Date	Time length of study	Scanner used	Initials/ Signature

Any signs of spillage and/or contamination, in the room used?

Yes		No	

All surfaces cleaned and disinfected?

Yes		No	

Scanner and coils cleaned and disinfected?

Yes		No	

Biological material stored/disposed appropriately?

Yes		No	

All contaminated consumables disposed appropriately?

Yes		No	

Comments:

Date	Time length of study	Scanner used	Initials/ Signature

Any signs of spillage and/or contamination, in the room used?

Yes		No	

All surfaces cleaned and disinfected?

Yes		No	

Scanner and coils cleaned and disinfected?

Yes		No	

Biological material stored/disposed appropriately?

Yes		No	

All contaminated consumables disposed appropriately?

Yes		No	

Comments:

Date	Time length of study	Scanner used	Initials/ Signature

Any signs of spillage and/or contamination, in the room used?

Yes		No	

All surfaces cleaned and disinfected?

Yes		No	

Scanner and coils cleaned and disinfected?

Yes		No	

Biological material stored/disposed appropriately?

Yes		No	

All contaminated consumables disposed appropriately?

Yes		No	

Comments:

Appendix 6. Protocols created

Author's Note: *The optimized MRI MSK protocols developed during this project can be requested directly to the Author, using the email indicated in the beginning of this Dissertation.*

The protocols are in a digital format that can only be read and used by in MRI scanners from the same manufacturers as the Scanner(s) used to create the protocol(s).

Appendix 7. Image quality Assessment Questionnaire

Image Quality Assessment Questionnaire

Thank you very much for accepting to collaborate on the present project which focuses on improving the overall image quality and scan time of all MRI examination currently performed at the NOC.

Please do not insert any information that may identify you as the person who answers this questionnaire. Only indicate your current work title.

On the following questionnaire you will be shown 15 clinical cases, each one with 2 groups of MRI images: A and B.

Both groups of images represent the same body structures and protocol yet the acquisitions parameters of each sequence, as well as others aspects of each may differ between the groups and within the group.

Please observe carefully all sets of images and answer the following questions accordingly, for each case, to the best of your abilities.

You are more than welcome to make commentaries and suggestions about the cases shown to you.

Please disregard any text information that you may find when viewing the images, as it might influence your judgment. Only focus on the quality of the images, please.

Case 1:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 2:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 3:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 4:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 5:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 6:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 7:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 8:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 9:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 10:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 11:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 12:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 13:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 14:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 15:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 16:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

Case 17:

1) Did you have any doubts about the image Weights shown to you? _____
    a. Which ones and why? _____
2) Do both groups of images have diagnostic quality? _____
    a. Which group would you prefer to use for reporting? _____
3) Comparatively which group of sequences have better image Contrast? _____
4) Which group as better Resolution? _____
    a. Is the increase in Resolution significant? _____
5) Which group as better SNR? _____
    a. Is the difference significant? _____
6) Do you see any artefacts? _____
    a. Which artefact in which sequence? _____
    b. Are they affecting the diagnostic quality of the images? _____
7) Is there any other sequence/plane that you consider to be better fitted to assess the case shown to you? _____

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