Contrast to Noise Ratio (CNR)

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Signal to Noise Ratio (SNR) is a concept familiar to most MRI radiographers. It reflects image quality, based on the relative brightness seen in our images as a consequence of actual detected signal (true patient anatomy) versus randomly superimposed (unwanted) signals. Better image quality comes as a result of a larger difference between these two, i.e. a higher SNR. As such, sequences are usually built and saved within our scanners to provide acceptable SNR for the clinical questions to be answered. If on occasion an image is produced that is lacking in SNR, radiographers should be able to readily and instinctively spot this and decide on how to best repeat the sequence to make an improvement – usually by either compromising the image resolution and/or the scan time in some way.
Contrast to Noise Ratio (CNR) however is understood less well. Whilst it also gives us a measure of image quality, it seeks to describe the ability to perceive neighbouring structures of differing tissue type, i.e. their difference in their SNRs.
In this way, the level of CNR to achieve diagnosis may be higher in some body areas where lesions are subtle, perhaps liver MRI, yet lower where inherent contrast is strong, for example angiography. So in short, images may be high or low in SNR and CNR independently of each other, and depending on what you are looking at, this may be just fine.

To help understand the concepts of CNR further, Melany Palmer, Senior MRI radiographer at the Royal Bournemouth Hospital, has investigated how to improve CNR, whilst at the same time recognising the inevitable and consequential trade-offs.

To improve the contrast to noise ratio (CNR), the difference in signal intensity between adjacent structures needs to be enhanced. This is achieved by exploiting the intrinsic and extrinsic contrast parameters to either enhance signal from the relevant tissues, or decrease the signal from the normal tissues. (Westbrook, Kaut Roth and Talbot, 2011).

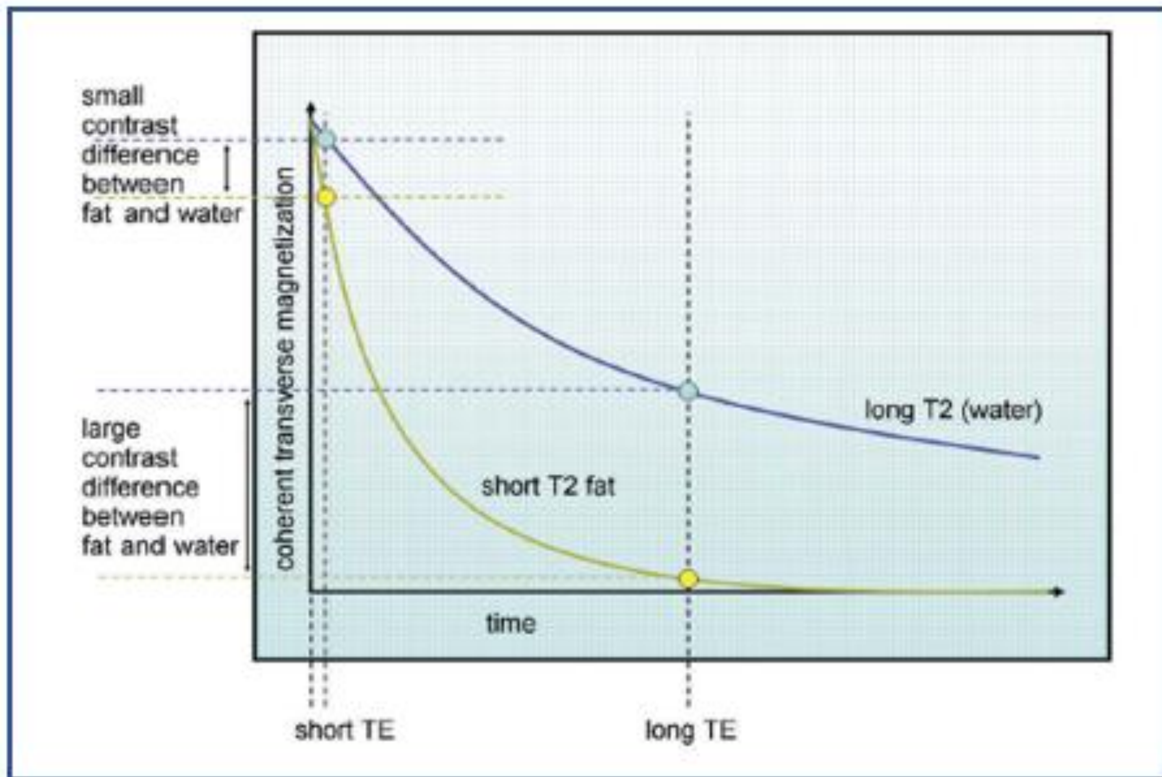

Figure 6.1. The T2 decay curves in fat and water (Westbrook, 2016, p.18).

Pathological tissues are typically associated with oedema or a high capillary density (McRobbie, et al., 2017). The use of T2-weighted images, which display water as hyperintense to the adjacent tissues, will therefore improve the CNR between pathology and the surrounding normal tissue. Figure 6.1 (Westbrook, 2016, p.18) displays the T2 decay curves of fat and water, demonstrating the large contrast difference achieved with the long TE values utilised in T2-weighted images. Due to water molecules being spaced further apart than fat molecules (Westbrook, Kaut Roth and Talbot, 2011), water has a longer T2 decay time than fat, as their spin-spin interactions are more infrequent than in tightly packed fat molecules. Their T2 time is therefore relatively long compared to fatty tissues, resulting in a larger contribution to the signal, thus appearing relatively hyperintense to the adjacent structures. If T2-weighted images are produced with conventional spin echo sequences, long TR and TE values are required which results in a long scan time (McRobbie, et al., 2017). FSE sequences are typically used to achieve T2-weighting as they significantly reduce scan times. The image contrast is however, compromised by the multiple RF pulses which reduce the effects of the spin-spin interactions in fatty tissue, thereby lengthening their T2 times. Fat is displayed as hyperintense relative to the

surrounding tissue, which may obscure any fluid content within the tissue.

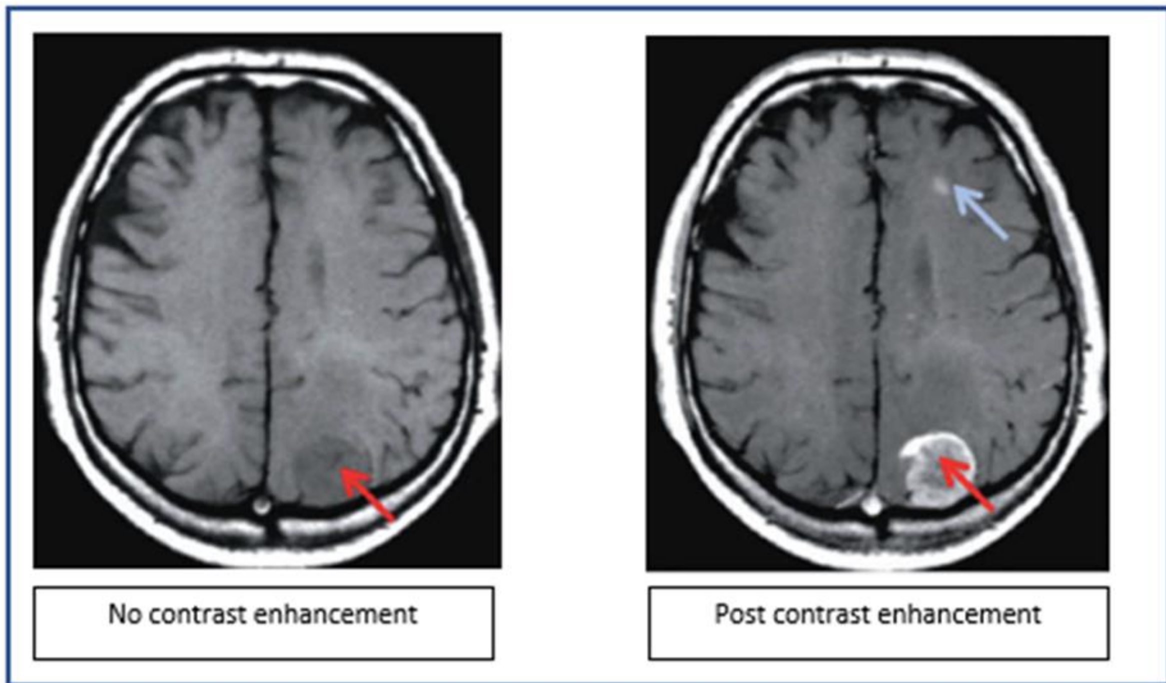

Figure 6.2. Axial T1-weighted image of the brain pre- and post-contrast enhancement, in a patient with metastatic disease (Westbrook, Kaut Roth and Talbot, 2011, p.377).

Some disease processes do not have a high-water content and will have a low CNR relative to the surrounding tissue (Westbrook, 2014). T1-weighted images produce a higher SNR than T2-weighted images; however, they generally display pathology and water as isointense. This is demonstrated in figure 6.2 (Westbrook, Kaut Roth and Talbot, 2011, p.377) where a brain metastasis is hypointense to the brain tissue (red arrow) prior to gadolinium contrast enhancement. Xiao, et al. (2016) ascertain a method to improve differentiation of these tissues, is to introduce a contrast agent. The majority of MRI contrast agents contain gadolinium as it possesses a high magnetic moment and is the most stable ion with unpaired electrons. This renders them strongly paramagnetic, thereby having positive magnetic susceptibilities which shortens the T1 relaxation time of the neighbouring water molecules, resulting in an increased signal intensity on T1-weighted images. The same metastasis is demonstrated post gadolinium contrast enhancement (red arrow), showing improved CNR. A small metastasis (blue arrow) that was inconspicuous without contrast enhancement, is

now visible on the post-contrast image. Dorazio, et al. (2014) note that pre- and post-contrast enhanced images are usually required for diagnosis, which adds to both scan time and cost.

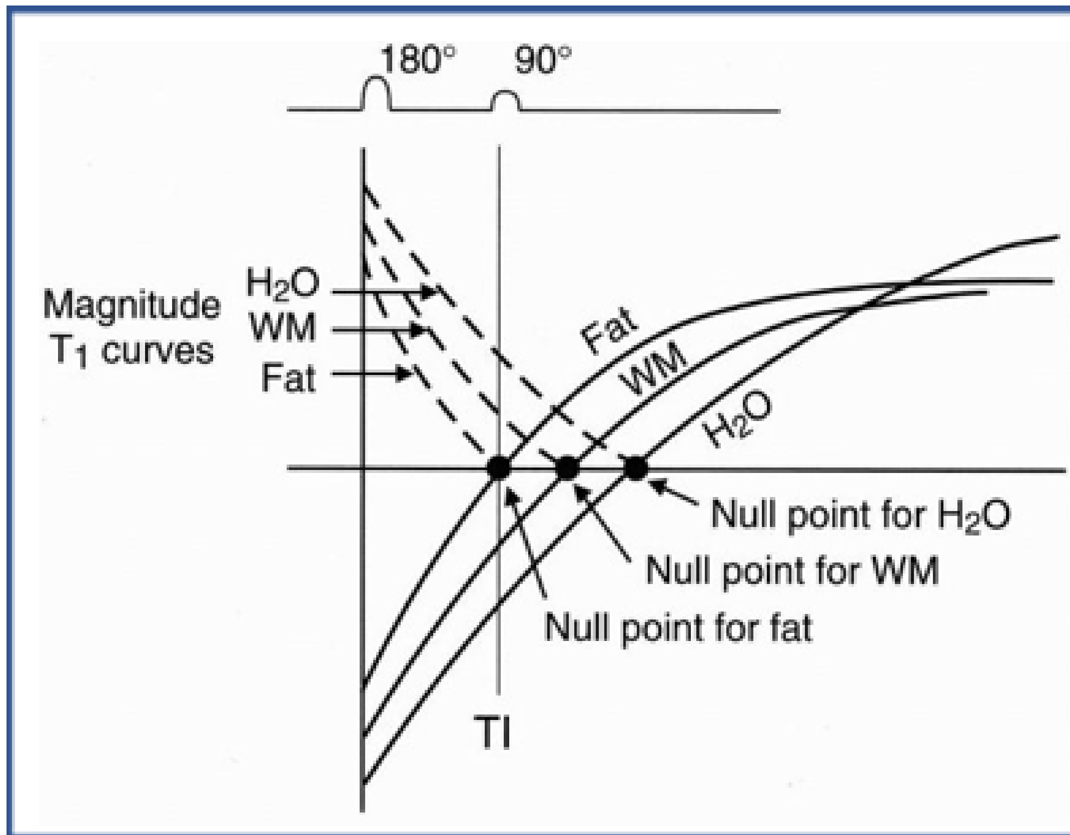

Figure 6.3. The different null points of fat, water (H_2O) and white matter (WM). In the STIR sequence, the TI is set so that the T1 recovery curve of fat crosses zero at the time of the 90° excitation pulse (Hashemi, Lisanti and Bradley, 2017, p.84)

Tissue suppression techniques are used to selectively suppress either fat or water signals (Hashemi, Lisanti and Bradley, 2017), which enhances the tissues of greater interest, such as pathology, thus improving the CNR. Figure 6.3 (Hashemi, Lisanti and Bradley, 2017, p.84) illustrates the inversion recovery (IR) pulse sequence which utilises a 180° inversion pulse at the beginning of the sequence to fully saturate the spins. The excitation pulse is applied at a time TI (time to inversion), which is set at the null point of fat in the diagram, so that only these spins will be fully saturated, and will therefore not contribute to the signal. The TI can be set to the null point of individual tissues depending on their T1 recovery times, thus determining the weighting of the image.

Short TI inversion recovery (STIR) sequences are used to supress fat signals and fluid attenuated inversion recovery sequences (FLAIR) are used to null fluid signals. IR sequences do however, require long TR values to allow for full T1 recovery, which increases the acquisition time. All tissues with a similar T1 time will be supressed at the same TI value and can therefore not be differentiated. For this reason, STIR sequences cannot be used post-gadolinium enhancement due to its T1 shortening effects which renders the pathologies T1 time equivalent to that of fat.

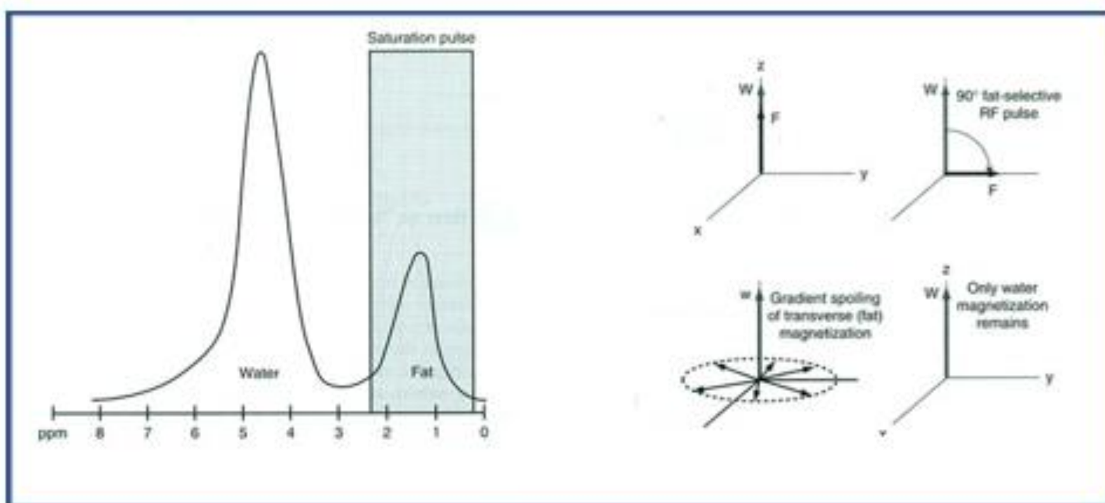

Figure 6.4. The fat peak is selectively saturated by a narrow-bandwidth RF pulse (Questions and Answers in MRI, 2017c).

The different precessional frequencies of fat and water can also be exploited to improve the CNR (Del Grande, 2014). At 1.5 T the precessional frequency of fat protons is 220 Hertz (Hz) lower than water protons. This allows for chemical shift selective suppression (CHESS) of either fat or water protons. CHESS is proportional to the main magnetic field; higher field strengths experience a wider shift between the fat and water peaks, allowing for more selective saturation, while lower fields may have heterogenous fat suppression, as the distance between the peaks is shortened, and overlap may occur. Figure 6.4 (Questions and Answers in MRI, 2017c) demonstrates fat suppression which utilises a 90° pre-saturation pulse with a narrow bandwidth centered on the resonant frequency of fat. The diagram further illustrates how the fat spins are flipped into the transverse plane and dephased with the use of a spoiler gradient, so that only magnetisation from water protons contribute to the signal.

Compared to IR sequences, CHESS techniques are relatively fast with a high SNR (Del Grande, 2014). They are however, sensitive to field inhomogeneities and are therefore less suited to large FOVs, off-centre imaging, and imaging of metallic implants, which increase susceptibility artefacts. STIR imaging in contrast, is insensitive to field heterogeneity and is therefore widely used in these instances.

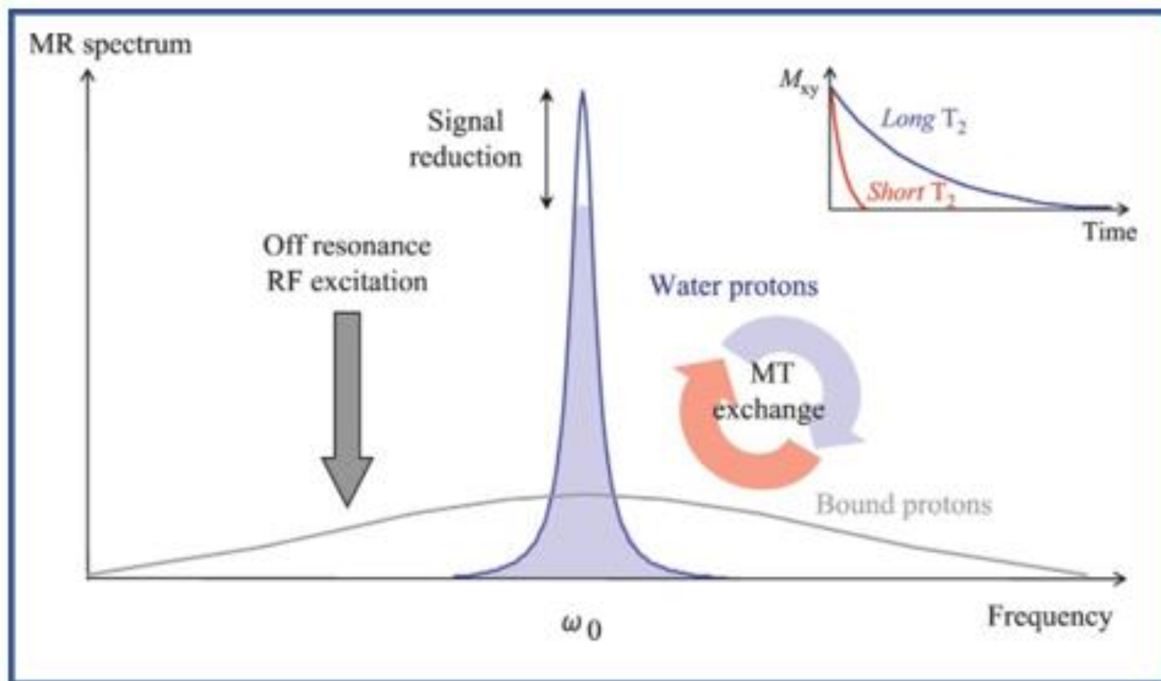

Figure 6.5. MT exchange between free water protons and bound water protons (McRobbie, et al., 2017, p.140).

Magnetisation transfer contrast (MTC) is a technique used to supress background tissues, thereby improving CNR by enhancing visualisation of smaller vessels and certain disease processes (Westbrook, 2014). Hydrogen protons within the tissues are typically classified into two groups; the “free pool” of mobile water molecules, and the “bound pool” of tightly bound macromolecules (Gambarota, 2012). The MR signal is usually generated from the free pool of molecules, as those in the bound pool have very short T2-relaxation times, thus contributing very little to the signal intensity. As demonstrated in figure 6.5 (McRobbie, et al., 2017, p.140), MTC is achieved by utilising an off-resonant RF pulse applied prior to the excitation pulse. This saturates the bound protons which causes them to exchange some of their saturated magnetisation to the free protons (Westbrook, 2014), resulting in reduced signal intensity from the protons in the free pool.