

Overview of Signal Detection and the RF Chain

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1. Introduction

This presentation is designed to give an overview of the building blocks of an MRI receive RF chain, starting with the local MRI coil going all the way to the image processor. Further information on this topic can be found in (1). This book contains a detailed introduction to MRI as well as other diagnostic imaging systems.

Figure 1 shows a typical receive chain consisting of the following building blocks:

- RF coils
- Detuning circuits
- Matching networks
- Low noise preamplifiers (LNAs)
- Mode Matrix and Eigencoil combiner circuitry
- Coil cables and Baluns
- Antenna selector
- Automatic level adjustments
- Receivers

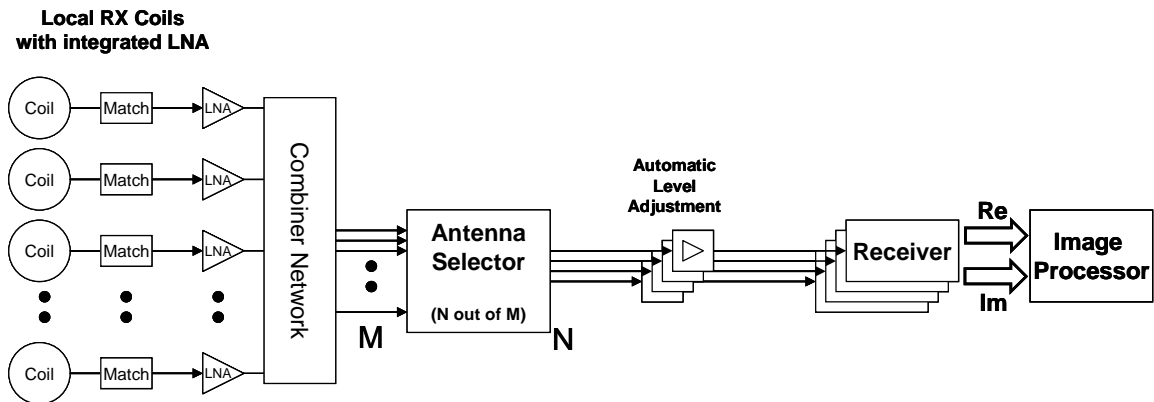


Figure 1: Building blocks of a typical RF receive chain.

The purpose of the RF chain can be summarized as follows:

“Select and extract a set of MRI signals from a subset of local MRI coils in a patient safe and user friendly manner, with high SNR and high linearity, and convert them from analog RF signals to digital complex k-space data.”

Some important parameters to characterize an RF chain are:

- Frequency
- Bandwidth
- Dynamic Range
- Noise figure
- Linearity
- Channel crosstalk
- Size of Antenna selector
- Local SAR

2. Building blocks of the RF chain

2.1. MRI coil including detuning, matching and LNA

Figure 2 shows an MRI coil in the vicinity of a conductive sample and the associated equivalent circuit (2-5). As can be seen, the coil can be modeled as inductor with losses originating from coil conductor, sample and radiation.

A well designed MRI coil exhibits a resistance which is mainly due to the sample and results in a high unloaded to loaded Q ratio.

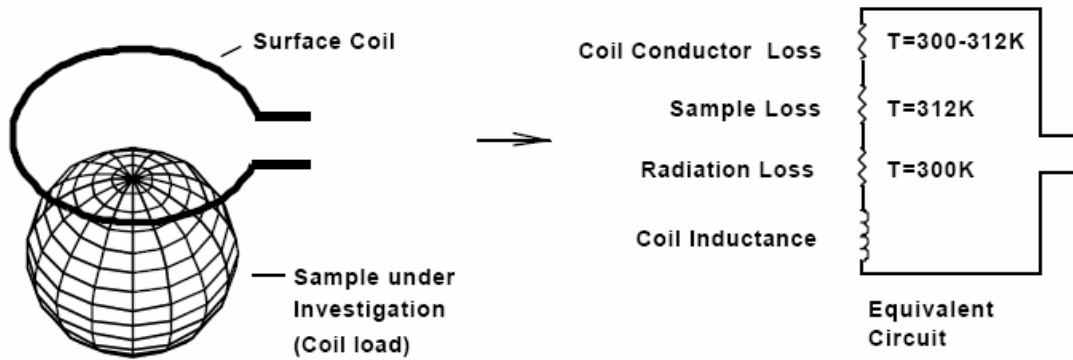


Figure 2: Equivalent circuit for a sample loaded local coil

In order to extract the MRI signal from the coil with highest possible SNR, the coil impedance has to be transformed to a impedance that insures a minimum noise figure of the connected low noise preamplifier (LNA). Typically, this noise match impedance is 50 Ohms.

Figure 3 shows the most important components of an MRI coil, which are detuning circuit, tuning and matching circuit, LNA and coil cable with one or more Baluns.

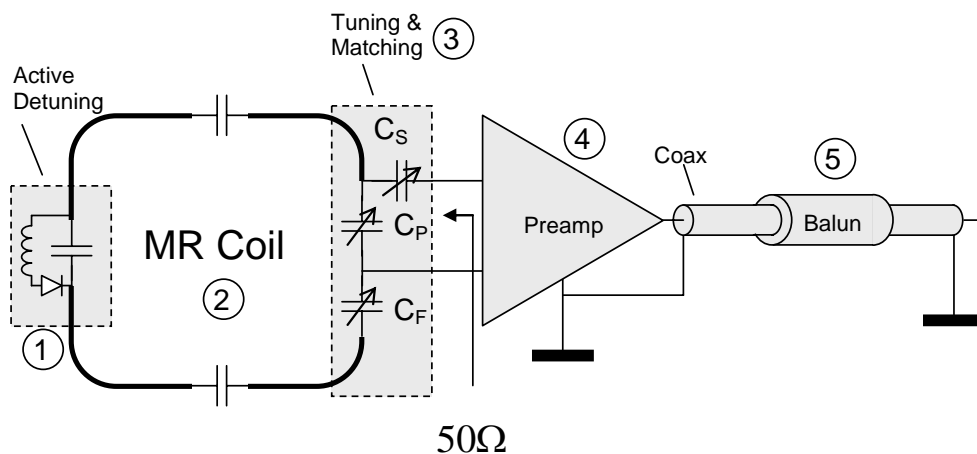


Figure 3: Local coil with detuning circuit, tuning and matching network, low noise preamp (LNA) and coaxial cable with Balun .

In array coil applications, the preferred method of matching coils to preamplifiers involves so called “isolating matching networks” (6). These networks serve two purposes: They transform the coil impedance to the desired noise match impedance (in this example 50 Ohms) and at the same time, they do transform the preamp input reactance into a high impedance at the coil feed point. The latter is important in order to reduce current flow on the coil and thus reduce the effect of inductive coupling on the coil input impedances.

Figure 4 shows such an isolating matching network including the formulas to calculate it.

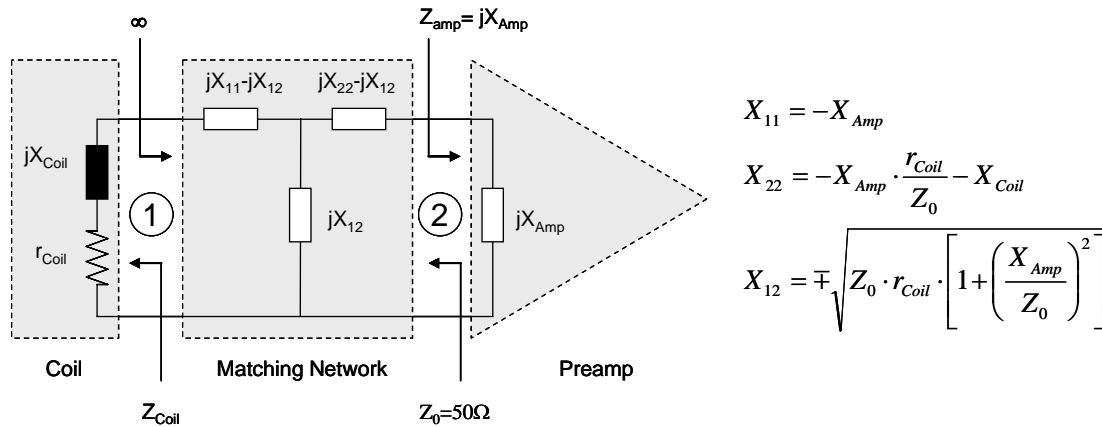


Figure 4: How to calculate the components of a isolating matching network from coil input impedance Z_{coil} , preamp input reactance X_{Amp} and noise match impedance Z_0 .

Modern preamplifiers for MRI purposes have a noise figure of less than 0.5dB, i.e. the preamp reduces the available SNR by no more than 5%. Also, such preamps exhibit a deviation from linearity of less than 1%. This means, that the maximum available MRI signal, which typically stems from 3D sequences with long TR, will stay below the 0.1dB compression point of the preamplifier.

2.2 Mode Matrix and Eigencoil combiner circuitry

While in the past array coils have been designed using LP or CP coil elements, the most recent developments are arrays using Mode Matrix or Eigencoil combiners (7, 8). These arrays consist of clusters of LP coil elements which are pre-combined to a set of mode signals. Figure 5 shows an example of a Mode Matrix combiner connected to a three element cluster. The original signals R, M, L are fed into the Mode Matrix combiner and transformed to three new signals P, S, T. Eq. 1 shows the Mode Matrix for the given example.

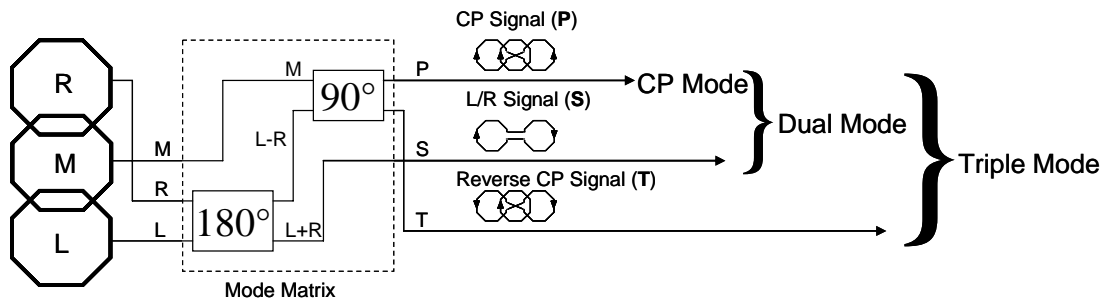


Figure 5: Example of a mode matrix for a three element coil cluster

$$\begin{bmatrix} P \\ S \\ T \end{bmatrix} = \underbrace{\begin{bmatrix} \frac{1}{2} & -\frac{j}{\sqrt{2}} & -\frac{1}{2} \\ \frac{1}{\sqrt{2}} & 0 & \frac{1}{\sqrt{2}} \\ \frac{1}{2} & \frac{j}{\sqrt{2}} & -\frac{1}{2} \end{bmatrix}}_{\text{Mode Matrix}} \cdot \begin{bmatrix} R \\ M \\ L \end{bmatrix} \quad [1]$$

The primary signal P is equivalent to the CP signal of a Loop-Butterfly combination. The two other signals contain array signal information from those imaging regions where the CP combination is not equal to the optimum array weighting.

For non-parallel imaging applications, the CP Mode signal contains sufficient SNR to achieve high image quality. If however parallel imaging techniques like SENSE, SMASH or GRAPPA are used, the additional signals from Dual- or Triple Mode deliver the spatial information needed for this type of image reconstruction (9-15). In order to improve SNR in the periphery of the image, Dual- and Triple Mode signals can also be used for non-parallel imaging.

When using all three mode signals P, S, T, the combined array SNR is the same as when using the original signals R, M, L. However, the new signals P, S, T offer more flexibility to the user: Each of the original signals covers only a part of the FOV, while the mode signals cover the entire FOV. With mode signals, it is thus possible to scale the number of receiver channels between one and three depending on the application.

2.3 Coil cables and Baluns

As has been indicated in Figure 3, one or several “Balun” (balanced-to-unbalanced) structures have to be incorporated with the coil cable.

These Balun structures minimize the flow of shield currents and thus prohibit the cable to act itself as antenna.

This is especially important during the sequence transmit phase, since excessive shield currents can lead to tissue heating and injuries. Therefore, care has to be taken to incorporate a sufficient number of these Baluns, also called “RF traps”, into the cable design and to make sure that these Baluns are properly tuned.

Figure 6 shows the two most common Balun structures used with MRI coils.

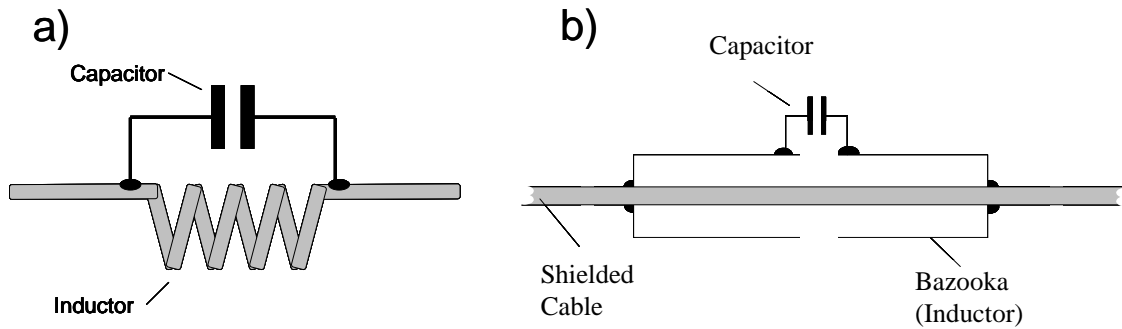


Figure 6: Two examples of Baluns: a) Balun with discrete inductance b) Bazooka Balun

2.4 Antenna selector

The antenna selector is a very important part of a modern workflow oriented MRI system. The antenna selector allows selecting a subset of the available MRI signals for further processing. The selection process is guided by two parameters:

1. Origin of Signal
Only those signals that come from within the region-of-interest (ROI) are selected
2. Mode of Operation
Defines how many mode signals from a given set of coils are selected.

Figure 7 shows a state-of-the-art antenna selector which allows selecting up to 32 signals out of 76.

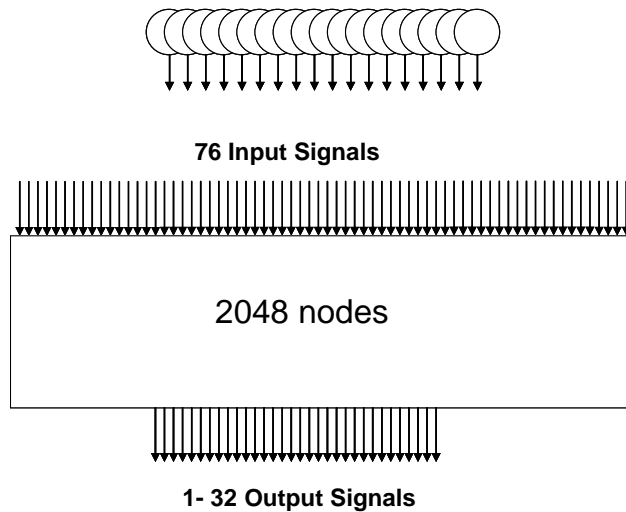


Figure 7: State-of-the-Art antenna selector (Siemens Magnetom Avanto) with 76 input signals and up to 32 selected receiver channels

2.5 Automatic level adjustment

The dynamic range (DR) of a signal is defined as the margin between the maximum peak power of the signal (e.g. $PS = -24\text{dBm}$ @ 1.5 T local coil LNA input) and the noise level ($PN/\text{Hz} = -174\text{ dBm/Hz}$ at room temperature)

The intrinsic dynamic range of the receive path must be considerably greater than the dynamic range of the signal in order not to sacrifice the signal-to-noise ratio (SNR), e.g. $> 180\text{ dB/Hz}$. Generally, the dynamic range of the Analog-Digital-Converter (ADC) is the limiting factor within the receive path. A state-of-the-art 14 bit-ADC with 10 MHz bandwidth only offers a dynamic range of approximately 153 dB/Hz. To overcome this restraint, automatic level adjustment must be employed: The present MR signal level is matched to the maximum level of the ADC by means of additional switchable gain elements or attenuators (Figure 8). Thus, the SNR of small MR signals will not be affected by the limited dynamic range of the ADC. State-of-the-art MR systems will automatically adjust the MR signal level by one or more of three different methods:

- a) Gain is fixed for the complete image acquisition time. The maximum MR level is forecasted based upon the coils used, sequence, slice thickness, and number of slices.
- b) Gain is fixed for certain echoes of the image. Usually, only a few k-space lines (within the center of k-space) have maximum level. All other echoes could be collected with a higher gain level.
- c) Gain is dynamically adjusted depending on the present MR signal level. This takes advantage of the fact that usually only the center portions of the centered k-space lines have the maximum level. Thus only a very small percentage of the k-space values are affected by the limited ADC dynamic range. The dynamic switching could be based on a forecast, or by a level detection circuit.

Methods b) and c) require careful calibration of the hardware in order to avoid image artifacts.

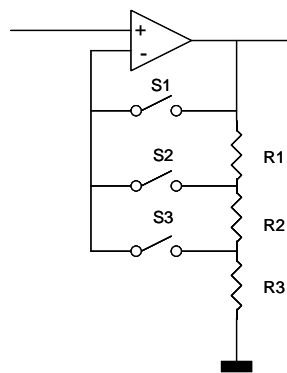


Figure 8: Basic principle of a switchable gain amplifier with several discrete gain settings

2.6 Receivers

The receiver performs a quadrature demodulation of the MR signal by multiplying the MR signal $u(t) = \text{Re}[F(t)]\cos \omega_0 t + \text{Im}[F(t)]\sin \omega_0 t$ with an in-phase LO ($\cos \omega_0 t$) and a quadrature LO ($\sin \omega_0 t$)

$$u_1 = u(t) \cos \omega_0 t = 1/2 \text{Re}[F(t)] + 1/2 \text{Re}[F(t)]\cos 2\omega_0 t - 1/2 \text{Im}[F(t)]\sin 2\omega_0 t \quad (4)$$

$$u_2 = u(t) \sin \omega_0 t = 1/2 \text{Im}[F(t)] - 1/2 \text{Re}[F(t)]\sin 2\omega_0 t + 1/2 \text{Im}[F(t)]\cos 2\omega_0 t \quad (5)$$

Afterwards terms with $2\omega_0$ are suppressed by a low pass filter (LPF). These filtered signals are representing the in-phase and quadrature components of the rotating frame signal $F(t)$. These signals are fed as a digital time series with time step T into the image processor. State-of-the-art receivers employ a fully digital demodulator (Fig. 9). The digitized signal $u(t)$ is simultaneously multiplied with digital $\cos \omega_0 t$ and $\sin \omega_0 t$ series, realized by a numerically controlled oscillator (NCO). The following digital LPF fulfills three tasks: Eliminating the $2\omega_0$ components, reducing the intrinsic (object) and artificial (noise figure of the Rx-path) noise, and reducing the noise of the ADC. Since the clock rates of commercially available ADCs do not permit direct conversion of one or more MR signals, one or more analog image rejection (IR) mixer stages must be employed (Figure 9). With their help, the received MR signal is translated from its carrier frequency f_0 to one or more intermediate frequencies f_{IF} . Afterwards, the signal is fed via an anti-alias LPF into the ADC.

If the intermediate frequency f_{IF} is chosen to be low enough, one ADC can sample signals from several channels in a TDMA (time division multiplex access) fashion.

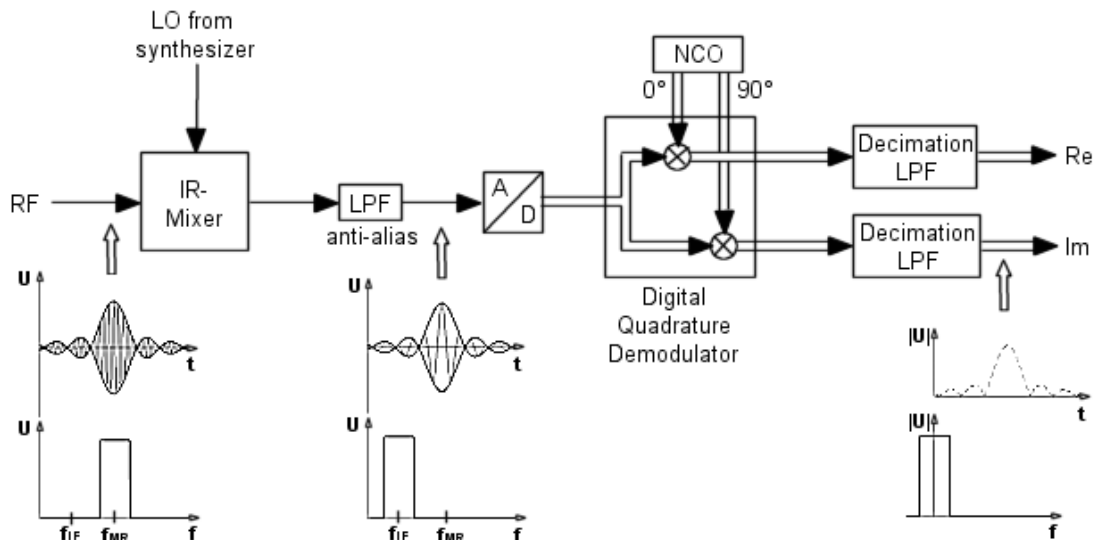


Figure 9: Block diagram of a state-of-the-art MR Receiver

3. Summary

The overall architecture of the RF receive chain should address workflow issues and modern trends like whole body and parallel imaging.

Careful design of every building block in the receive chain is necessary to insure highest signal integrity as well as patient safety.

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