BUILDING AN ARRAY OF SIX INDUCTIVE DECOUPLED RECEIVE COILS FOR MRI

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Building an Array of Six Inductive Decoupled Receive Coils for MRI

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Abstract

In research project, it was examined if an array of six receive coils for MRI has a higher SNR than an array of two receive coils with the same surface. In order to do so, both an array of six and an array of two coils were built. The coils were impedance matched and tuned at a frequency of 298.10 MHz. Q-spoiling circuits were used to detune the coils during the transmission of the RF pulses. To decouple the coils, the method of inductive decoupling was used. To compare the two arrays, low tip angle gradient echo images of phantom bottles were acquired. The conclusion of these images was that an array of six receive coils for MRI indeed had a higher SNR than an array of two receive coils with the same surface.

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INTRODUCTION

In the field of MRI, there are various devices used to acquire an image. At first, there is the main coil, in this case, a 7 Tesla MRI scanner located at the ‘Leiden University Medical Centrer’. Then, there are coils to transmit a signal, the RF pulse, to the human body. Because of that signal, a signal is transmitted by the human body itself, or to be more precise, by the rotating nuclei of the hydrogen atoms in the human body. To detect this signal, receive coils are used, and that is where this research project was about. The goal of this project was to examine if an array of six receive coils has a higher SNR than an array of two coils. The total surface of the two arrays is the same. The hypothesis is that an array of six coils will have a higher SNR, because the amount of signal increases due to the higher number of coils, but the amount of thermal noise coming from the examined tissue will not.

However, to build this array of six receive coils, several steps had to be taken. For example, the coils had to resonate at the same frequency as the hydrogen nuclei, namely 298.10 MHz. In order to achieve this, tuning of the coil was necessary. Furthermore, the amount of signal loss had to be reduced to a minimum in order to obtain a high SNR. This is called matching of the coils and capacitors were used to do this. One of the most important parts of the coils are the decoupling circuits. They are used to make sure that the coils only detect signal coming from directly beneath them, and not signal coming from tissue beneath other coils. At last, Q-spoiling circuits were used to make sure the coils do not resonate at 298.10 MHz during the transmission of the RF pulse.

In the upcoming sections, a detailed explanation of this building process is given. Chapter 1 describes the basics of magnetic resonance imaging. So it is explained how the MRI scanner is used to acquire images and how it is possible to tell the difference between, for example, fat and water. In chapter 2, the properties of the Smith Chart are described and in the following 2 chapters the usefulness of the Smith Chart to characterize the coil is explained. Chapter 3 is about matching the receive coils which is used to reduce the amount of signal loss at the transition between coil and coax cable. In the following chapter, chapter 4, tuning of the coils in described. The fifth chapter describes how a Q-spoiling circuit is built and where it is used for. Chapter 6 gives a mathematical explanation of the coupling between the coils and shows the importance of the decoupling circuits. Chapter 7 describes the actual building process of the coils.

The last chapter, chapter 8, contains the data and results of this project and in addition, it contains the conclusion of the project. In short, it is concluded that an array of six coils indeed has a higher SNR than an array of two coils.

1. MRI BASICS

1.1 Applying an External Magnetic Field

The process of making a MRI scan is based on the magnetic properties of several elements present in the human body. In this section, a brief explanation of this process will be given. Let's take a look at the element hydrogen, which is one of the elements with magnetic properties found in the human body. In fact, there is an abundance of hydrogen atoms in the human body, for example in water and fat, which makes it a good element to detect by using the MRI scanner. The nucleus of the hydrogen atom is a rotating proton. Because of that, it has a precessing magnetic spin moment.

In normal conditions, so without an external magnetic field, the magnetic spin moments of all hydrogen nuclei in the human body are directed randomly like in figure 1a, so the net magnetization of all atoms is zero. This changes when an external B-field, for example from a MRI scanner, is applied. Is this research project, a Philips Achieva whole body scanner was used [1]. The field strength of this scanner is 7 Tesla, which is over 200000 times the earth's magnetic field strength at the equator. The strong magnetic field, directed in the positive z-direction, then forces the magnetic spin moments to align with or against the field, see figure 1b.

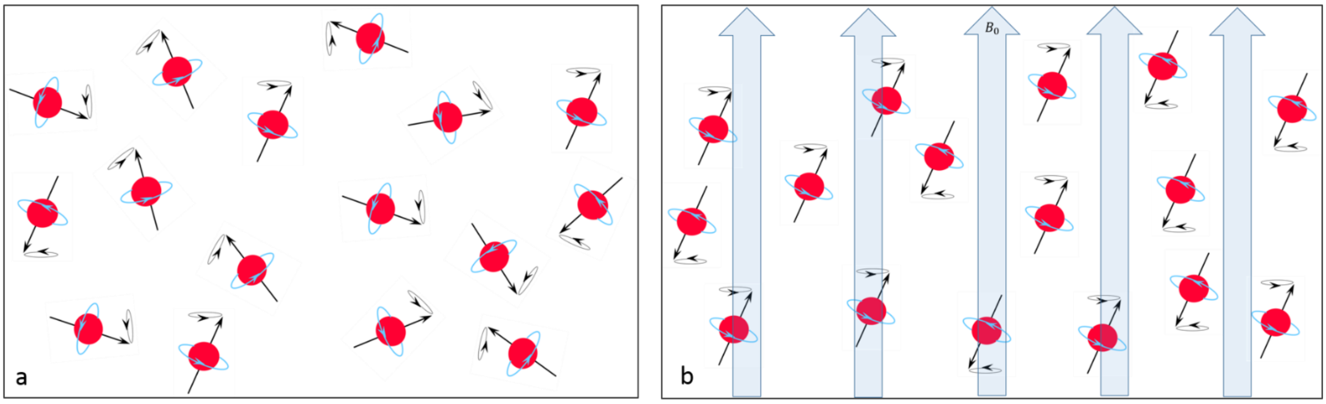


Figure 1
a: Direction of magnetic spin moments of hydrogen atoms when no external B-field ($\boldsymbol{B_0}$) is applied
b: Direction of magnetic spin moment of hydrogen atoms when an external B-field ($\boldsymbol{B_0}$) is applied

For quantum mechanical reasons, aligning with the field is a lower energy state, so more hydrogen atoms will be in that state after the external field is applied [2]. Just as in the normal conditions, the net magnetic field in the x- and y-direction is cancelled out because the hydrogen nuclei are not rotating in phase. But in this situation, the net magnetization in the z-direction will not be cancelled out because more hydrogen nuclei are aligned with the field. This results in a net magnetic field, called M, in the positive z-direction, as is shown in figure 2 [3].

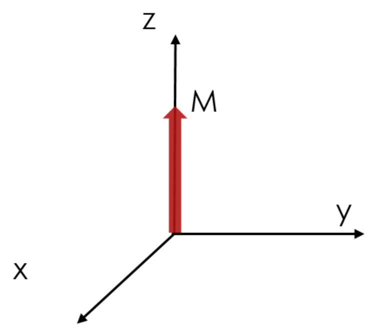


Figure 2: Net magnetization (M) after an external B-field is applied

1.2 The RF Pulse

But this net magnetization is not enough to be able to produce images. This is because M is static and therefore it is not measurable by using a coil, and it is overruled by the external B-field applied by the scanner, which is in the same direction (figure 3a). So two things are needed to make imaging possible: change the direction of M and make it non-static. Both things can be accomplished at the same time by using a Radio Frequency (RF) pulse. When such a pulse is applied when someone is in the MRI scanner, the net magnetization starts rotating. In the case of figure 3b, the rotation is in the ZY-plane. The angle of rotation depends on the duration of the RF pulse. The longer the pulse is applied, the greater the angle of rotation. In figure 3, the consequences of a 90° Radio Frequency pulse are shown. Figure 3c shows the situation after the pulse, when M starts rotating in the XY-plane. The frequency of this rotation is given by equation 1:

$$\omega_0 = \gamma B_0 \tag{1}$$

Where ω_0 is called the Larmor frequency, γ is the gyromagnetic ratio and B_0 is the strength of the external magnetic field [4]. The gyromagnetic ratio of hydrogen is 42.58 MHz/T [5].

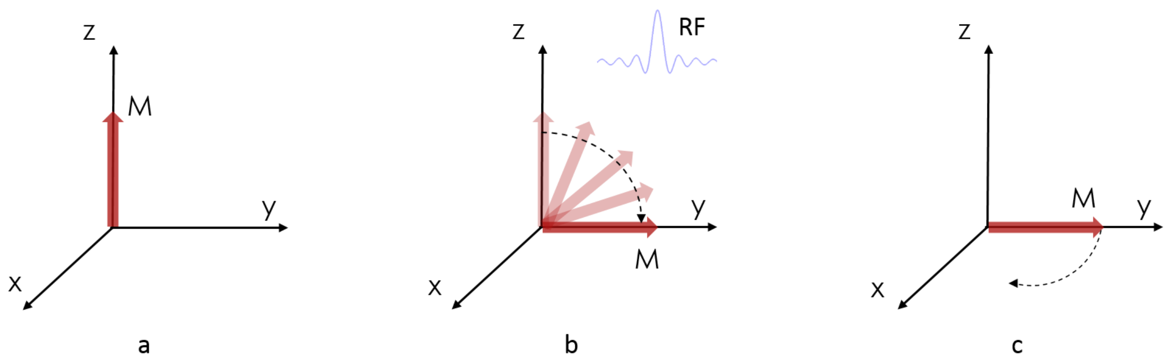


Figure 3
a: Net magnetization (M) after the external B-field is applied
b: During the RF pulse (blue line), the net magnetization (M) starts rotating towards the XY-plane
c: When a 90° pulse is applied, M has components in the XY-plane only and rotates in this plane

So that means that in a 7 tesla field, M oscillates with a frequency of 298.1 MHz. Such an oscillating magnetic field is detectable by using a coil. This is because an electrical current is induced in the coil as a result of the rotating net magnetic field. The voltage, U, of this induced current can be calculated by using Lenz law (equation 2):

$$U = -N\frac{d\Phi}{dt} \tag{2}$$

Where N is amount of windings of the coil, Φ is the magnetic flux and t is the time. So the value of the voltage, U, tells something about the amount of hydrogen atoms present near the measuring detection coil. Naturally, the more hydrogen atoms there are, the higher the voltage U.

1.3 Relaxation

But knowing the amount of hydrogen atoms present in a certain voxel (volume element) is not all of the work, because hydrogen is found in many types of tissue, of which water and fat are the two that are most important. So a method to distinguish whether a hydrogen atom is in water or in fat is needed. A way to make this distinction is by measuring the T_1 value. But before an explanation about T_1 is given, it is useful to know what happens directly after the RF pulse is turned off. When the RF pulse is turned off, the magnetic field in the z-direction, B_0, forces the magnetic spin moments to align with or against B_0 again. This is called relaxation or longitudinal magnetization recovery. This happens not all at once, but takes some time. The time it takes for 63.33 % of the magnetic spin moment to return to its direction before the RF pulse, is called T_1 (figure 4) [6]. This time, T_1, is different for different types of tissue. So for that reason, making the distinction between two types of tissue, for example fat and water, is possible by using the relaxation time T_1.

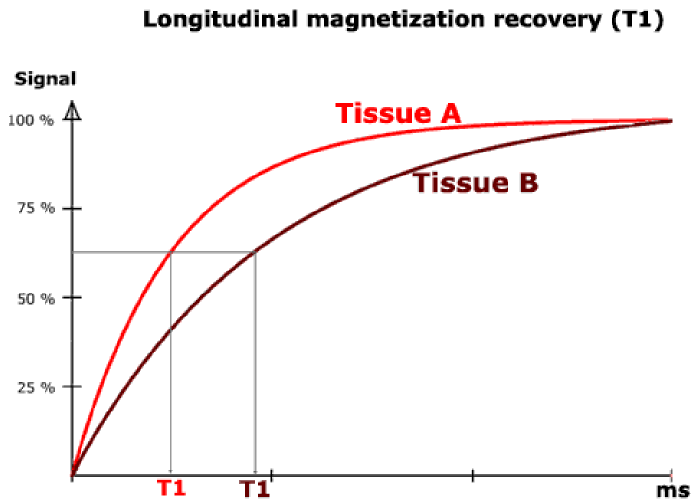


Figure 4: Graph showing the longitudinal magnetization recovery of 2 different tissues, A and B. For both tissues, the value of T_1 is indicated.

2. SMITH CHART

The Smith Chart is a coordinate system which can be used to tune and match receive coils used for making MRI scans. The Smith Chart was developed by P.H. Smith in the 19th century and shows, for example, the impedance of a coil. The impedance can be seen as a complex extension of the resistance. Where the resistance only tells something about the magnitude, the impedance tells both about the magnitude and the phase angle of an Alternating Current. So the resistance is the real part of the impedance, which is given by equation 3:

$$Z = |Z|e^{i\theta} = R + iX \tag{3}$$

Where Z is the impedance, |Z| is the voltage (V) divided by the current (I) [7], i is the imaginary unit, θ is the difference in phase between V and I, R is the resistance and X is called the reactance. The reactance consists of an inductive part (X_L) and a capacitive part (X_C), which is shown in equation 4.

$$X = X_L - X_C = \omega L - \frac{1}{\omega C} \tag{4}$$

Where ω is the frequency. So in the end, in terms of R, L and C, the impedance is given by equation 5:

$$Z = R + i\omega L - \frac{i}{\omega C} = R + i\omega L + \frac{1}{i\omega C} \tag{5}$$

Now, in a Smith Chart, the magnitude of the real part, which is the resistance, is represented by an infinite amount of circles. So every point on such a circle corresponds with the same value of R [8]. Some of these circles are shown in figure 5a, including the two most important ones, R = 0 Ω and R = 50 Ω.

Those circles are important because the former circle forms the outside of the Smith Charts and the latter crosses the middle of the Smith Chart corresponding to Z = 50 Ω for the coil, which means that it is impedance matched. More about matching will be told in further sections. The imaginary part of the Smith Chart is represented by an infinite amount of arcs, as is shown in figure 5b [8].

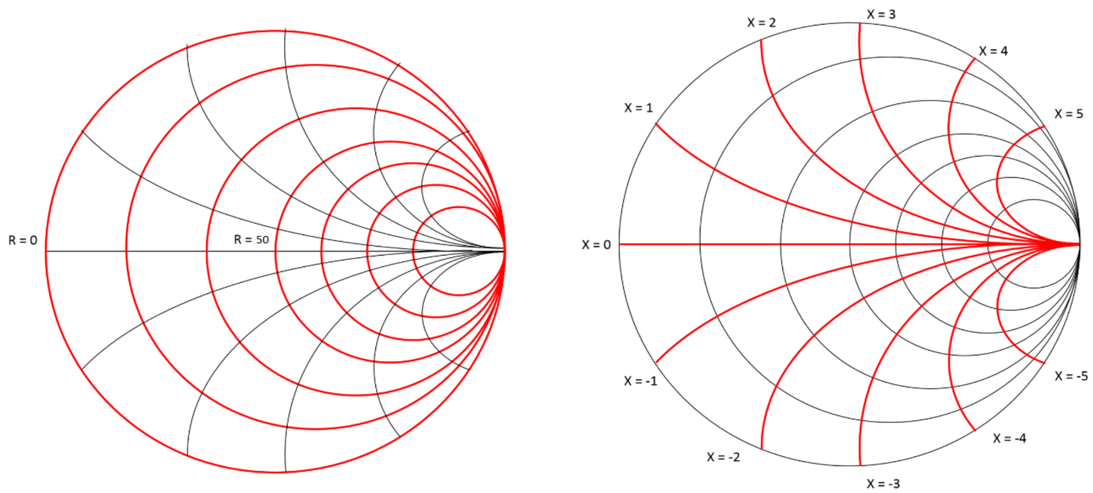


Figure 5
a: The real part of the smith chart consists of an infinite amount of circles, of which some are marked red
b: The imaginary part of the smith chart consists of an infinite amount of arcs, of which some are marked red

There are three important values of Z according to some special situations. The first one is Z = 0. In this case the complex resistance is zero, which means that, in the case of a coil, it is shorted. If the value of the impedance is infinite, it means that the coil is open [9]. The last important value is the earlier mentioned value Z = 50 Ω. This means that the coil is impedance matched resulting in no loss of signal due to reflections in the coax cable. Where these three values are on the Smith Chart is shown in figure 6.

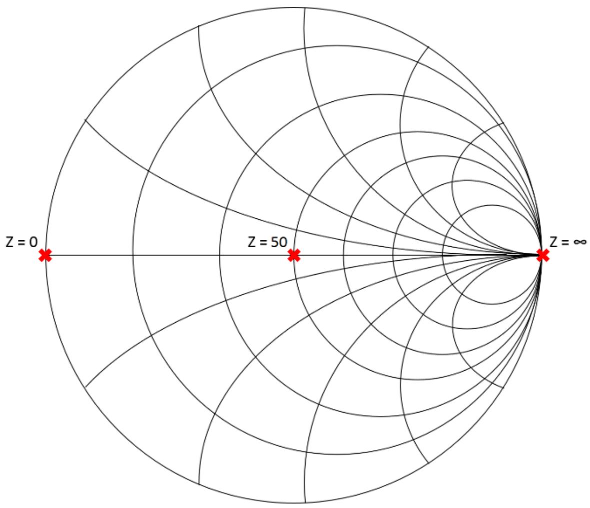


Figure 6: Three important values of Z. Z = 0 corresponds to a shorted coil, Z = ∞ to an open one and a coil with Z = 50 is impedance matched.

3. IMPEDANCE MATCHING

Matching of the coil, shortly mentioned in earlier sections, is used to reduce the reflection of signal at the transition between the coil and the coax cable. The coax cable, just like all devices used in the field of MRI, is built in such a way that it has an impedance of 50 Ω. As seen before, this value corresponds to the centre of the Smith Chart. The matching circuit used in this research project is shown in figure 7 and consists of two matching capacitors which are not part of the main coil. One of them is located between the main coil and the centre core of the coax cable [10]. The second capacitor is placed between the main coil and the ground of the coax cable.

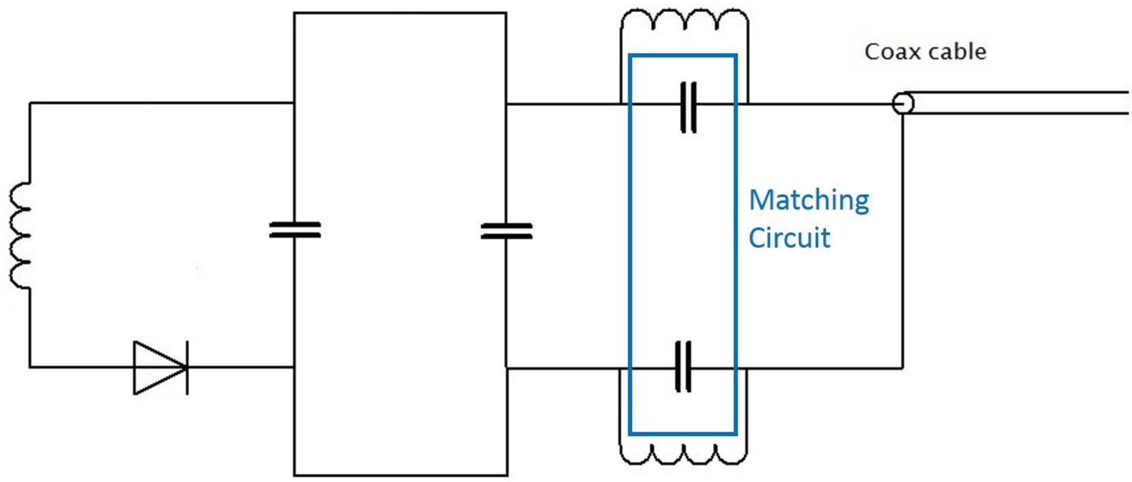


Figure 7: Matching circuit (inside blue box) consisting of two capacitors.

Because of the small size of the coil, small C values, approximately between 2 and 10 pF, were used to match the coil. As a result of this, calculating the needed values of C was very difficult because, for example, even the soldering material had a substantial contribution to the total impedance of the coil. Therefore, a trial and error way of matching was used. In this method, a Smith Chart was very useful because it could tell if capacitance had to be added or removed. For example, after soldering on a capacitor, the Smith Chart in figure 8 was obtained. In this figure, point A corresponds to the impedance value at ω = 298.10 MHz. It is clear that this coil is not matched in a proper way, since point A is not located at the centre of the Smith Chart. The first problem is the value of X, the imaginary part in

$$Z = |Z|e^{i\theta} = R + iX$$

This value has to be zero, but is positive in this case. Now movement in the counter-clockwise direction along the circles in the Smith Chart is needed in order to obtain X = 0 at ω = 298.10 MHz.

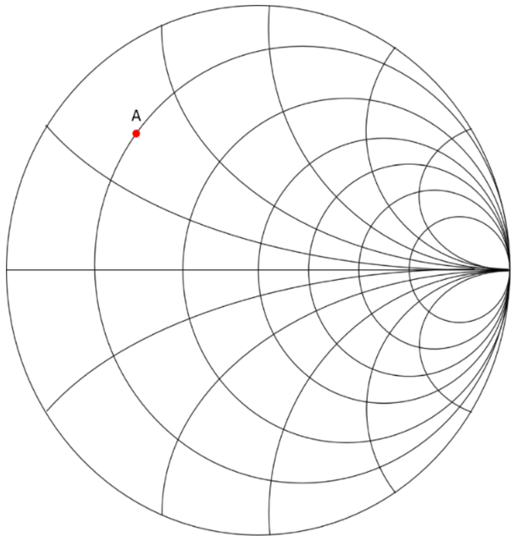


Figure 8: Example of a smith chart after soldering on a matching capacitor. Point A corresponds with the impedance value at a frequency of 298.10 MHz

To accomplish this 'movement', capacitance has to be added [11]. The reason for this can be seen in equation 4 of the Smith Chart section. Namely, the $1/\omega C$ term has a negative contribution to X, so if X is too high, adding capacitance will move point A along the circle in the counter-clockwise direction to point A'. If the correct value of C is used, A will move exactly to the real axis of the Smith Chart. This is shown in figure 9.

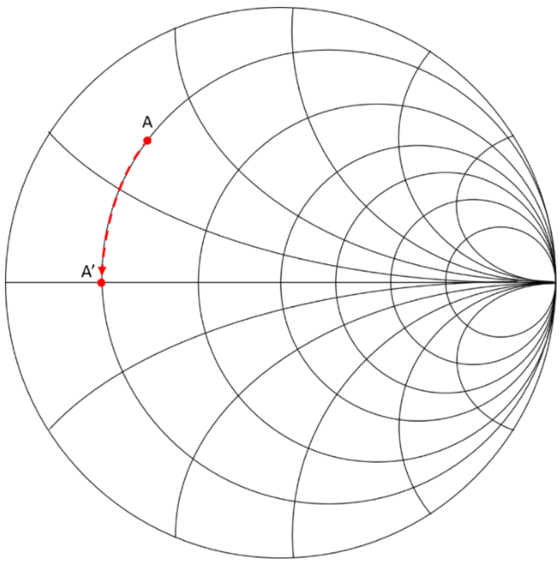


Figure 9: Movement of point A along the circle of the smith chart in the counter-clockwise direction after soldering on a capacitor with higher C.

Now, the X term in Z is zero, but A’ is not at the centre of the Smith Chart yet. To accomplish this, two things can be done. At first, the resistance could be increased by adding a resistor to the coil. In this case R would increase, which results in a movement of A’ along the centre line of the Smith Chart towards the middle, as is shown in figure 10. If the correct value of R is used, A’ will move exactly to the centre of the Smith Chart, A”, which means that the coil is matched in a proper way.

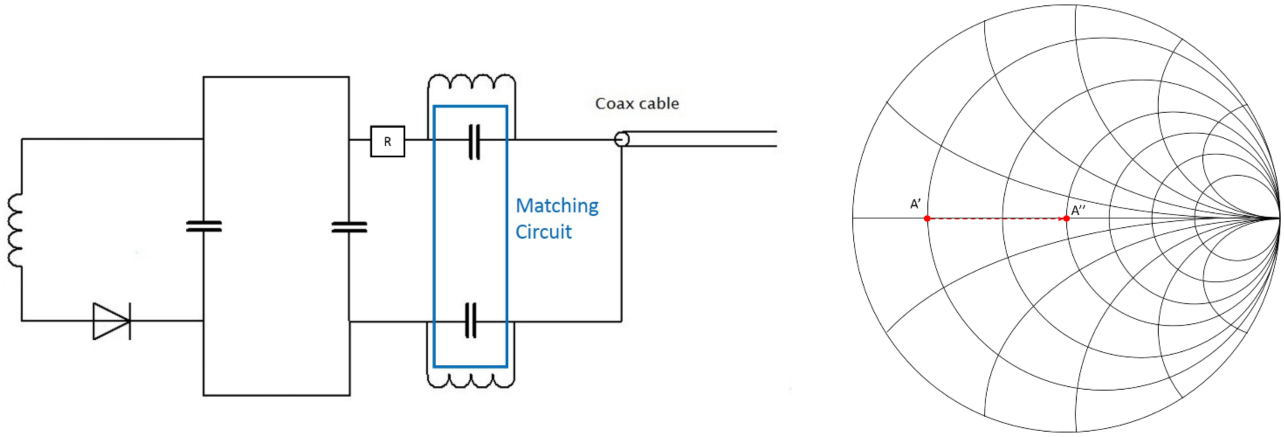


Figure 10
left: Coil after soldering on a resistor R used to match the coil
right: Movement over the central line of the smith after soldering on a resistor R.

However, in this research project, the second way to move along the resistance axis was used. In this method, changing the values of the tuning capacitors was used to match the coils. This method will be discussed in the tuning section.

4. TUNING

To make sure that the used receive coil resonates at the same frequency as the hydrogen nuclei in the body, tuning of the coil is necessary. So in the case of a 7 Tesla scanner, the coil has to be tuned in such a way that it resonates at a frequency of 298.10 MHz.

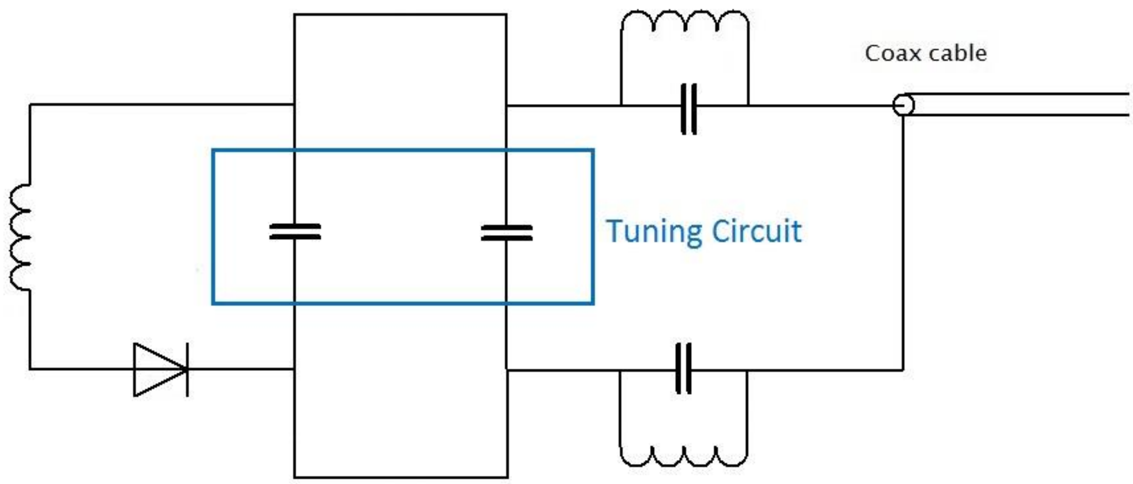


Figure 11: Tuning circuit (inside blue box) consisting of two capacitors.

Just like a matching circuit, a tuning circuit consists of two capacitors, only in the case of a tuning circuit, the capacitors are located in the main coil, see figure 11. For the same reason why a trial and error method was used to match the coil, it was used to tune coil. After two capacitors were soldered on the main coil as is shown in figure 11, a measurement with the Network Analyser was used to determine at which frequency the coil is resonating. In this case a s11 measurement was used. This means that the Network Analyser sends alternating currents in a large frequency range, in this case from 200 MHz to 400 MHz through the coil and then measures the signal coming back at every frequency. The ratio between the power of the ingoing or reference signal (P_0) and the measured power of the signal (P) is given on a decibel scale and can be calculated by using equation 6 [12].

$$G_{dB} = 10\ log_{10}\left(\frac{P}{P_0}\right) \tag{6}$$

A graph of such a s11 measurement is shown in figure 12.

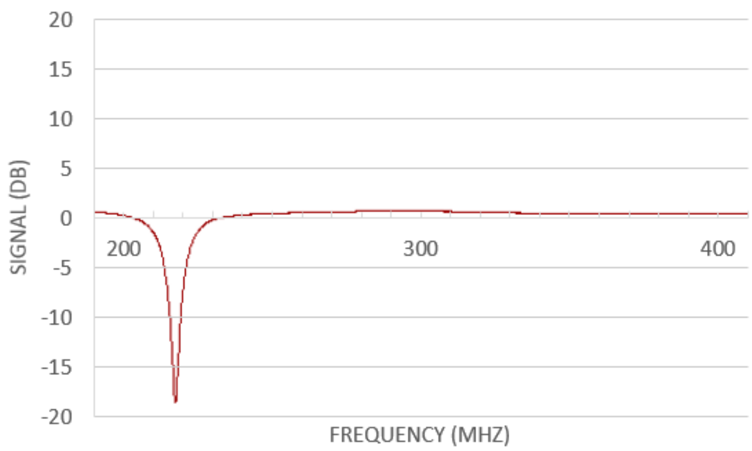


Figure 12: Picture of a s11 measurement made by a Network Analyser

So the negative peak shows at which frequency the coil resonates, because at that frequency the measured power is much lower than the reference power. Figure 12 shows a measurement of a coil which is not tuned in a proper way, this is because the peak is around 220 MHz instead of 298.10 MHz. To determine if the capacitance should be increased or decreased, equation 7 can be used.

$$\omega_{res} = \frac{1}{\sqrt{LC}} \tag{7}$$

Where ω_{res} is the resonance frequency of the coil, L is the inductance and C is the capacitance. So if, as in the case of figure 12, the resonance frequency is too low, the capacitance has to be decreased in order to increase ω_{res}. If the correct value of C is added, the resonance peak will shift and a graph like in figure 13 is obtained, in which the coil is tuned in a proper way.

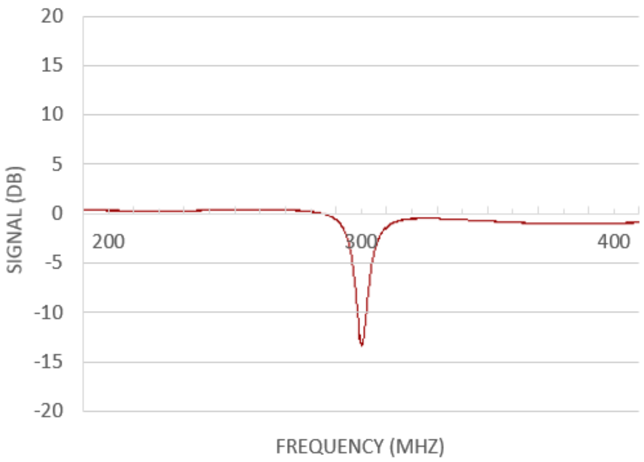


Figure13: Picture of a s11 measurement made by a Network Analyser

5. Q-SPOILING

The Q-spoiling circuit is an important part of a receive coil in MRI. It is used in receive-only coils because it works during the transmit RF pulse. During this pulse, the Q-spoiling circuit actively detunes the coil to make sure it does not resonate at 298.10 MHz. The reason for this detuning is that the RF pulse has a frequency of 298.10 MHz, just as the hydrogen atoms and the tuned and matched receive coil. So if the coil would not be detuned it would pick up the RF pulse signal. If that happens amplifiers and other devices attached to the coil can be damaged. The Q-spoiling circuit, shown in figure 14, consists of a capacitor, an inductor and a PIN-diode. It is important that the PIN-diode has no magnetic parts, because if it does the magnetic field produced by the superconducting magnet in the MRI scanner will be influenced [13]. The values of L and C depend on the amount of neighbours that a coil has. This is because the capacitor used in the Q-spoiling circuit is also a part of the main coil. A main coil with 2 neighbours has 2 inductive decoupling circuits, which will be mentioned later, and therefore a higher L than a coil with only 1 neighbour having 1 decoupling circuit. Because ${}^{1}/_{\sqrt{LC}} = 298.1\ MHz$ holds for both coils, a coil with 2 neighbours has a lower C value than a coil with 1 neighbour. On the other hand, ${}^{1}/_{\sqrt{LC}} = 298.1\ MHz$ holds for the Q-spoiling circuit as well, so a Q-spoiling circuit connected to a main coil with 2 neighbours has a higher L than a Q-spoiling circuit connected to a main coil with 1 neighbour. The working of the Q-spoiling circuit is based on the fact that direct current is stopped by a capacitor but not by an inductor. During the RF-pulse, the amplifier attached to the coil sends a direct current trough the inductors in the coil. In figure 14, this current is indicated by red arrows.

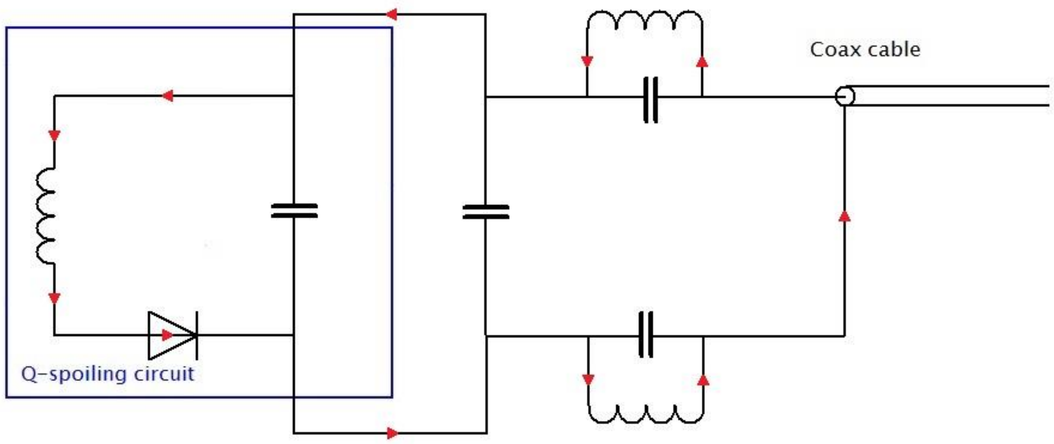


Figure 14: In the blue square there is a Q-spoiling circuit consisting of a capacitor, an inductor and a PIN-diode. Different values of C and L were used in different coils. The red arrows indicate the current going through the coil during the RF pulse.

As a result of the current, the PIN diode in the Q-spoiling circuit opens. When the PIN diode is open, the Q-spoiling circuit can be seen as a parallel LC circuit. In a parallel LC circuit, the impedance (Z) can be calculated by using equation 8.

$$Z(\omega) = -i\frac{\omega L}{\omega^2 LC-1} \tag{8}$$

Where ω is the frequency, i is the imaginary unit, and L and C are respectively the inductance and capacitance of the parallel LC circuit. If the correct values of L and C are chosen, the impedance will be ∞ at a frequency of 298.10 MHz. This is the case when $\omega = \frac{1}{\sqrt{LC}}$. As a result of this infinite impedance, the coil will not be able to pick up any signal coming from the RF pulse. Figure 15 shows a picture of a Q-spoiling circuit used in this research project.

Figure 15: Q-spoiling circuit attached to a main coil with 2 neighbouring coils used in the research project. Here L = 33 nH and C = 5.6 pF

6. COUPLING & DECOUPLING

6.1 Mutual Inductance

Because an array of multiple coils was used in this research project, mutual inductance has to be taken into account. Mutual inductance is a phenomenon which occurs when two coils are brought close to each other. For example, if two coils are placed next to each other and a current goes through one of them, the situation in figure 16 is obtained [14]. The current I_1 induces a magnetic field, B_1. Because coil 1 and 2 are close to each other, field lines of B_1 will cross the surface of coil 2. The crossing of these field lines is called the magnetic flux, ϕ_{21}.

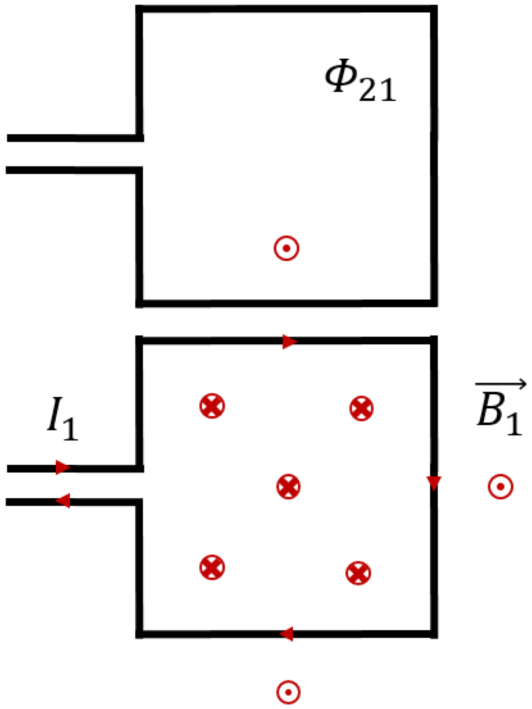


Figure 16: Two coils placed next to each other. A current goes through coil 1 and induces a magnetic field B_1. ϕ_{21} is the magnetic flux through coil 2 as a result of I_1.

The change of ϕ_{21} depends on the change of I_1 in the following way (equation 9):

$$\frac{d\Phi_{21}}{dt} = M_{21}\frac{dI_1}{dt} \tag{9}$$

Where ϕ_{21} is the magnetic flux through coil 2 as a result of I_1, M_{21} is called the mutual inductance and I_1 is the current through coil 1.

6.2 Coupling

Because of the mutual inductance between the multiple coils in the array, the signal of, for example, coil 1 will not be measured by the signal detector of coil 1 only. In the case of two coils, the voltage in coil 1 and coil 2 can be given by equations 10 and 11 [15]. In this equations, V_1 and V_2 are the induced voltages in the coils, i is the imaginary unit, ω is the frequency, I_1 and I_2 are the currents in the coils and M_{12} and M_{21} are the mutual inductances between coil 1 and 2.

$$V_1 = i\omega L I_1 + i\omega M_{12} I_2 \tag{10}$$

$$V_2 = i\omega L I_2 + i\omega M_{21} I_1 \tag{11}$$

As can be seen from equation 10 and 11, the voltage of coil 1 depends on both I_1 and I_2. So in the case of a receive coil array for MRI, this means that coil 1 is imaging what is below coil 1 and coil 2. This is called coupling and has a negative effect on the quality of the obtained images. A possibility to decrease M_{12} and M_{21} would be of course to increase the distance between the coils. But in that case the field of view (FOV) would increase just as the amount noise, so a lower signal to noise ratio will be obtained. Therefore, a different way to decrease M_{12} and M_{21} has to be used. One where the distance between the coils is unchanged.

6.3 Decoupling

A way to decrease M_{12} and M_{21} is by overlapping coils which are located next to each other [16]. An example of such overlapping coils is shown in figure 17.

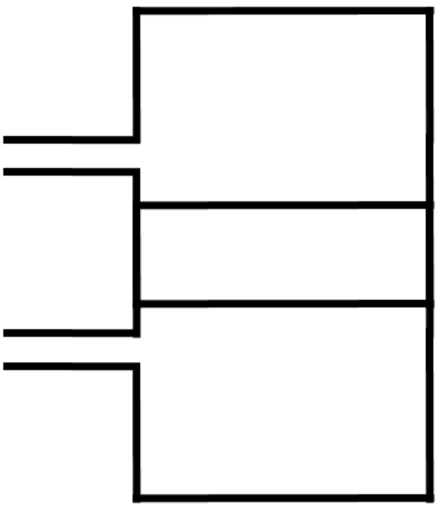


Figure 17: Two overlapping receive coils for MRI

Because of the overlap between coil 1 and coil 2, M_{12} and M_{21} will decrease. Let's take a look at, for example, the lower coil, coil 1. When a current is going through coil 1, a magnetic field will be induced. As a result of this field, a magnetic flux, ϕ_{21}, occurs. In the case of non-overlapping coils, ϕ_{21} is nonzero and therefore there will be mutual inductance between coil 1 and coil 2. But if the coils overlap in a proper way, the magnetic flux of the non-overlapping part of the coils will cancel out the magnetic flux of the overlapping part. See figure 18.

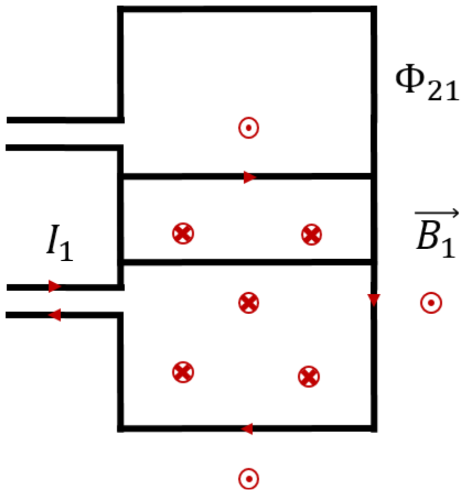


Figure 18: Two overlapping coils placed next to each other. A current goes through coil 1 and induces a magnetic field B_1. ϕ_{21} is the magnetic flux through coil 2 as a result of I_1.

So, if ϕ_{21} is zero, M_{12} and M_{21} will be zero as well. If that is the case, equation 10 and 11 will reduce to equation 12 and 13.

$$V_1 = i\omega L I_1 \tag{12}$$

$$V_2 = i\omega L I_2 \tag{13}$$

Now, the voltages in coil 1 and coil 2 depend on respectively the current trough coil 1 and the current through coil 2. The advantage of overlapping coils as a decoupling method is that the FOV decreases but the amount noise is unchanged, so the SNR increases. A disadvantage of overlapping coils is that the amount of overlap is not easily adjustable when the coils are made on PCB. So for that reason, another decoupling method was used during this research project, namely inductive decoupling.

In the case of inductive decoupling, loops of copper wire are used to build circular inductive decoupling circuits. The physical mechanism in this method is the same as for overlapping coils. By placing the decoupling circuits on coil 1 and coil 2 in the way shown in figure 19, overlap between the two coils is created. If the circuits are placed and shaped in a proper way, ϕ_{21}, and therefore, M_{12} and M_{21}, become zero. The advantage of this kind of decoupling is that the decoupling circuits are easily adjustable and therefore small changes in ϕ_{21} can be made.

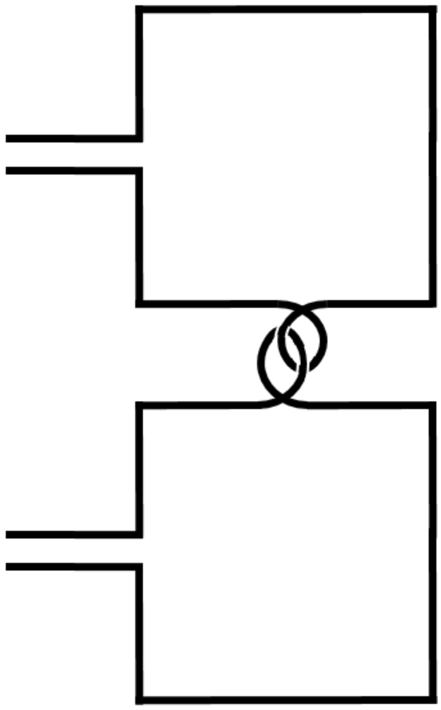


Figure 19: Two coils with inductive decoupling circuits.

7. COIL DESIGN

7.1 PCB Etching

All coils used in this research project were built on PCB's, printed circuit boards. To create the outlines of the coils on the PCB, a laser cutter was used to cut them in etching tape. After that, the coils were immersed in a chemical bath to remove unwanted copper not covered by etching tape [17]. See figure 20.

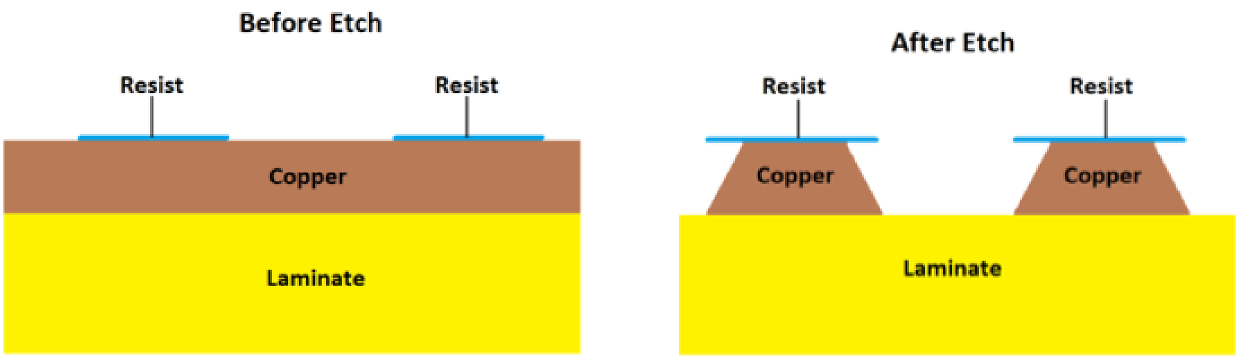
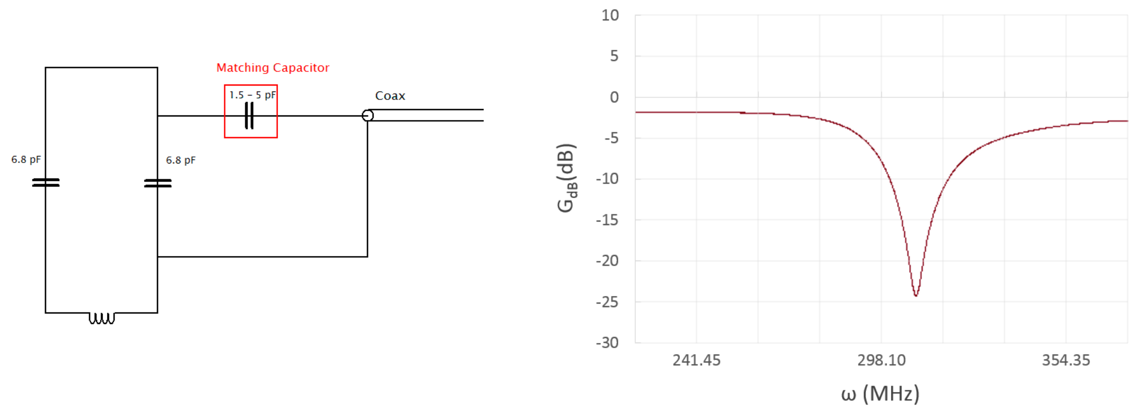


Figure 20: Situation before and after a coil was put in a chemical bath. In this case, the resist is etching tape cut in the correct shape by using a laser cutter.

7.2 Building a Single Coil

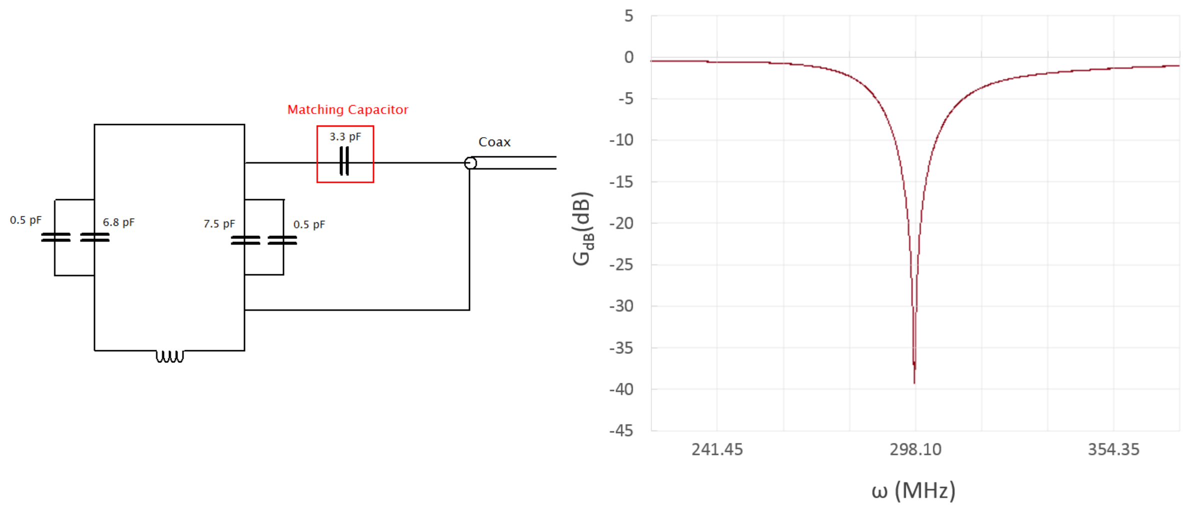
At first, the size of the coil had to be determined. The coils cannot be too small, because the value of the B-field in the region of interest is proportional to the size of the coil. On the other hand, when the coils are too big, the thermal noise coming from the region of interest will be too large and the signal to noise ratio will be lowered [18]. In this case, the first coil had a length and width of 40 millimeters. To determine which values of C should be used to tune and match the coils, the coil was built without a Q-spoiling circuit. Instead of the Q-spoiling circuit, a decoupling inductor had to be placed, because it is part of the main coil and therefore influences the resonance frequency. After building the coil shown in figure 21a, the resonance frequency was measured by using the network analyzer. See figure 21b.



(a) (b)

Figure 21: (a) Single coil consisting of two tuning capacitors, both with C = 6.8 pF, a matching capacitor between 1.5 and 5 pF and a decoupling inductor which is made by hand and has one winding.
(b) Result of a measured made on the coil in (a) by using a network analyser.

It can be seen that the values of the two tuning capacitors are too low, because the coil is not tuned at 298.10 MHz. So higher values of C were used to tune the coil at 298.10 MHz. Besides that, the variable matching capacitor was replaced by a fixed one with a value of 3.3 pF (figure 22a). ω_{res} was measured by using the network analyzer of which the result is shown in figure 22b.



(a) (b)

Figure 22: (a) Single coil consisting of two tuning capacitors, one with C = 0.5 + 6.8 pF and another with C = 0.5 + 7.5 pF, a matching capacitor with a value of 3.3 pF and a decoupling inductor which is made by hand and has one winding.
(b) Result of a measured made on the coil in (a) by using a network analyser.

7.3 Adding more coils

When more coils are added to the array, besides matching and tuning, another thing should be taken into account, namely the coupling between all the coils. At first, all coils where equipped with a Q-spoiling circuit. Q-spoiling circuits of coils with one neighbor consist of an inductor with L = 27 nH and a capacitor with C = 6.8 pF. If a coil has two neighbors, the Q-spoiling circuit consists of a capacitor with C = 5.6 pF and an inductor with L = 33 nH. This difference is due to the different amount of decoupling circuits on coils with one or two neighboring coils. So a coil with two neighbors has 2 decoupling circuits, and a coil with one neighbor has only one decoupling circuit. These circuits consist of a handmade inductor with one winding, and two of these circuits were twisted together as in shown in figure 23.

Figure 23: Decoupling circuit consisting of 1 inductor with 1 winding per coil twisted together. The decoupling circuits are made of silver wire.

At first, the decoupling circuits were made of silver wire coated with an isolating material (used in figure 23), but later on coated copper wire was used. The reason for this is the flexibility of the two materials. The copper wire itself and its coating are a little thinner than the silver wire, so the copper was easier to bend, and therefore smaller changes is the shape of the decoupling circuits were easier to make. The shape of the decoupling circuits determines the amount of signal of neighboring coils measured by a certain coil. The array shown in figure 24 was built and al coils were tuned and matched.

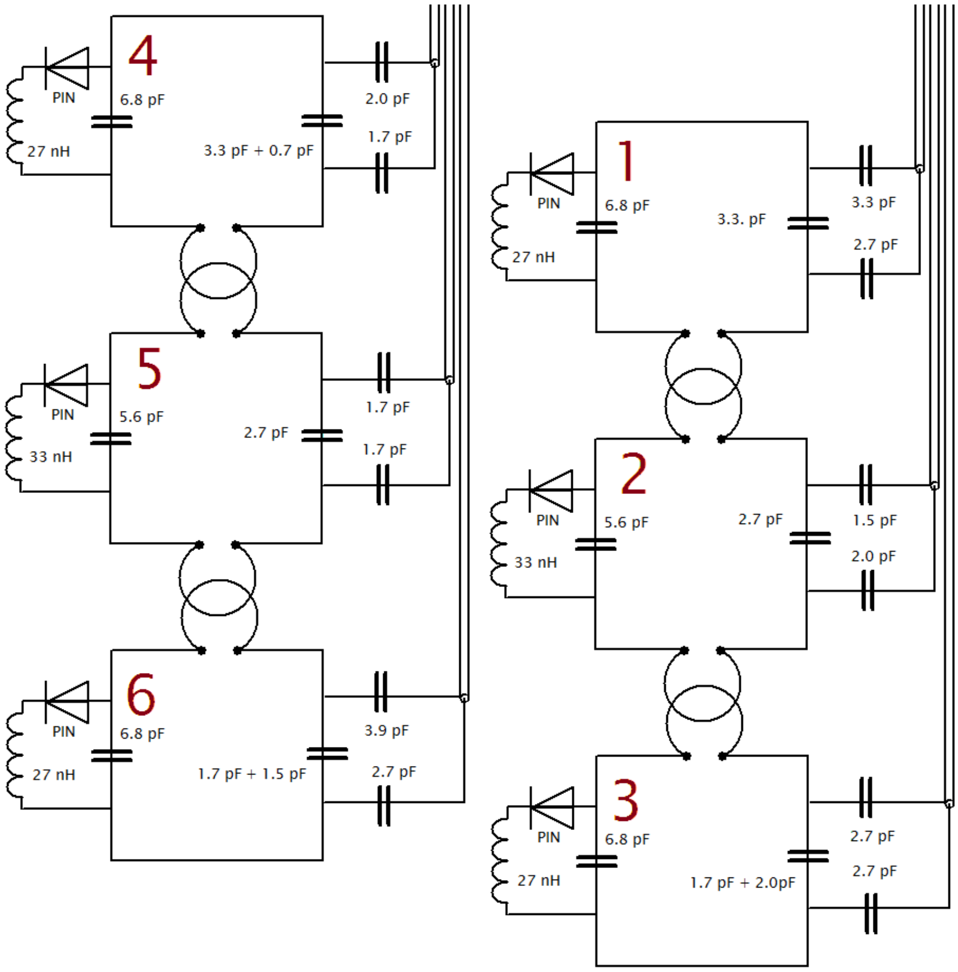


Figure 24: Array of 6 receive coils including matching, tuning, decoupling and Q-spoiling circuits. For the matching circuits, values of C between 1.5 and 3.9 pF were used. Tuning circuits had values of C between 2.7 and 6.8 pF. All decoupling circuits were made out of handmade single winding inductors twisted together as in figure 23.

It can be seen that different values of C were used to tune and match the different coils. There are two reasons for that. The first one has to do with the amount of neighbors a coil has and has been explained earlier. The second reason is that the shape of each decoupling inductor is slightly different because they are made by hand. Because of that different shape, each inductor has a different value of L and therefore different values of C were used to be able to tune all coils at a frequency of 298.10 MHz. After the tuning of the coils, the matching capacitors were soldered on and the decoupling circuits were shaped in the correct way. The matching and decoupling of two neighboring coils could be checked by doing a s11, s22 and s12 measurement at the same time. In figure 25, the Smith Chart in the upper left corner shows the matching of coil 2. Indicator >1 shows the impedance at a frequency of 298.10 MHz. The indicator in the upper right Smith Chart shows the impedance of coil 3 at a frequency of 298.10 MHz. The lower graph is the result of a s12 measurement.

In this case, during such a measurement, the network analyzer sends alternating currents with different frequencies through coil 2 and then measures the signal at each frequency through coil 3. In that way, the coupling between coil 2 and coil 3 at ω = 298.10 MHz can be measured. In figure 25 it can be seen that the coupling between coil 2 and 3 is -21.784 dB. This means that 0.6631% of the power of the signal with ω = 298.10 MHz sent through coil 2 is measured by coil 3 [19].

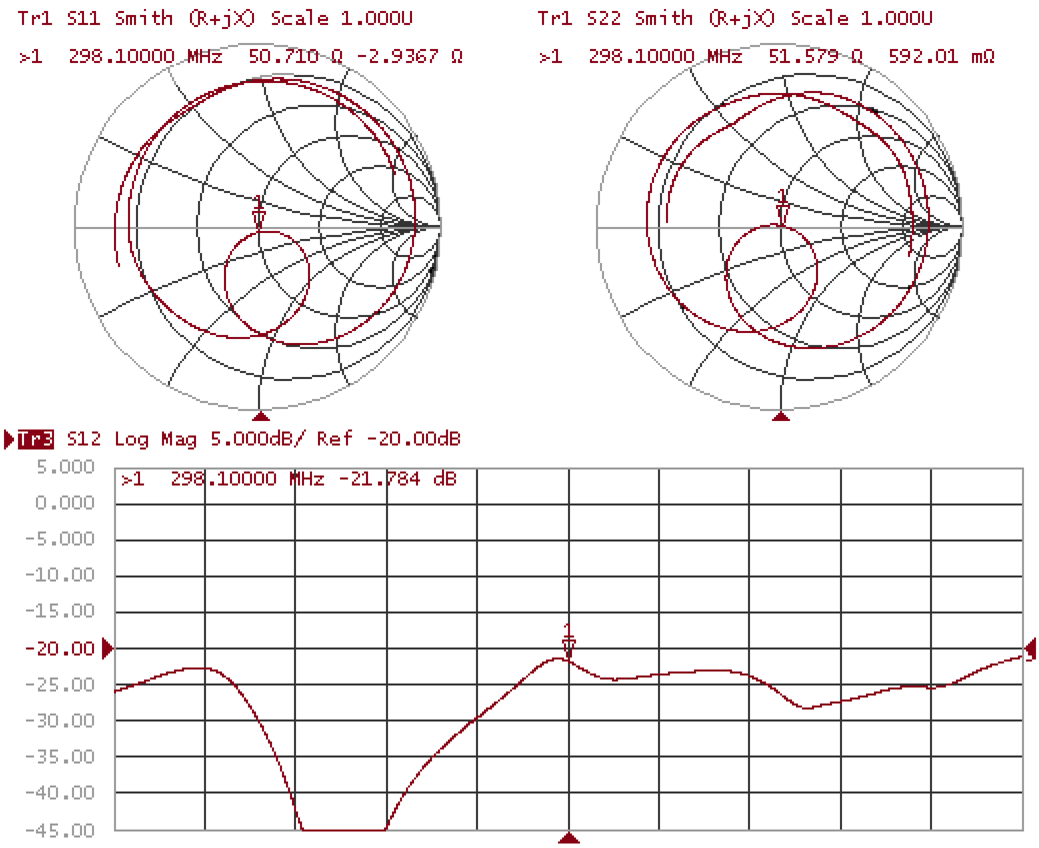


Figure 25: The upper left smith chart shows the result of a s11 measurement of coil 2 in the array of 6 shown in figure 5. The indicator >1 shows the impedance of coil 2 at ω = 298.10 MHz. The upper right smith chart shows the result of a s22 measurement of coil 3 in the array of 6 shown in figure 24. The indicator >1 shows the impedance of coil 3 at ω = 298.10 MHz. The lower graph shows the coupling between coil 2 and 3 determined by doing a s12 measurement. The indicator >1 shows the coupling at a frequency of 298.10 MHz.

By using s12 measurements, the coupling of each couple of coils was determined. The result of all these measurements is shown in figure 26. Figure 27 shows a picture of the completed coil used in this research project.

1	2	3	4	5	6	Coil
	-13,5	-22,5	-29,8	-18,4	-27,7	1
		-21,8	-25,5	-21,8	-15,3	2
			-41,4	-24,2	-13,7	3
				-13,8	-20,2	4
					-18,5	5
						6

Figure 26: Coupling of all coils used in the research project. The values are in dB and were determined by doing a s12 measurement with the network analyser.

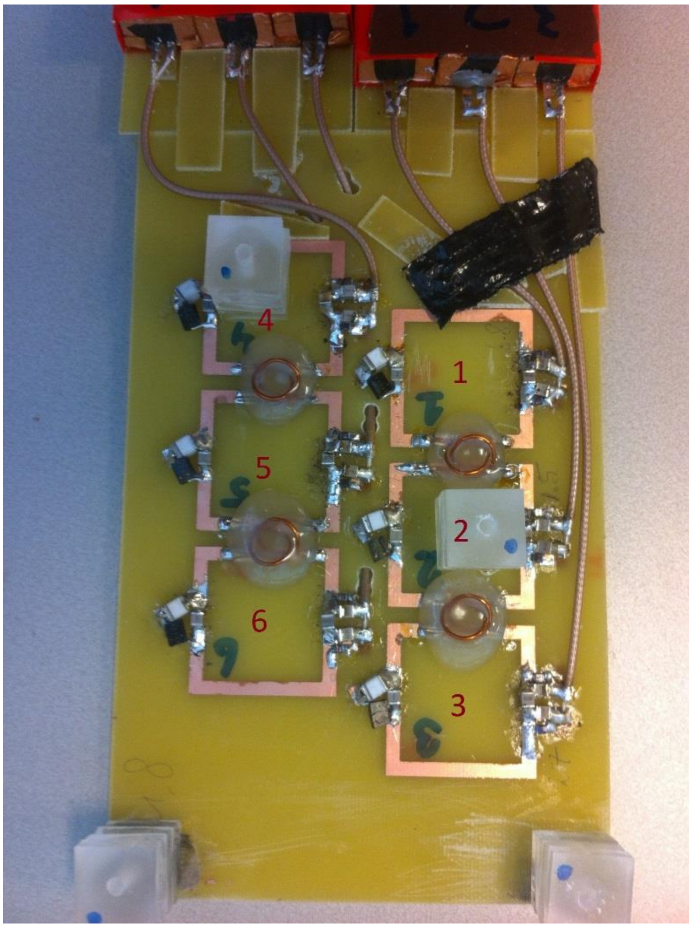


Figure 27: picture of the coil shown in figure 24

7.4 Cable Traps

To finish the array of receive coils, cable traps had to be added. Cable traps are electrical circuits used to eliminate the influence of the coax cable. This is because the cables have a substantial contribution to the impedance of the coil if no cable traps are used. When the cables move, the contribution to Z changes which influences the matching and tuning of the coil. To stop this, circuits as shown in figure 28 were placed between the coils and the coax cables. These circuits reduce the contribution of the cables to a minimum.

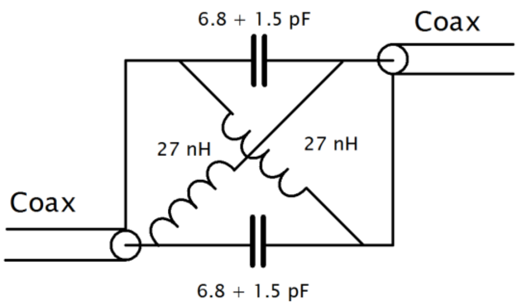


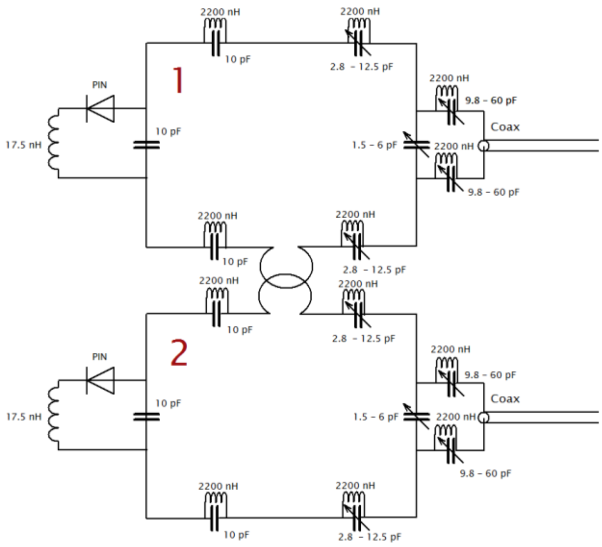
Figure 28: Cable trap used to reduce the contribution of a coax cable to the mathing and tuning of a receive coil for MRI

8. DATA & RESULTS

8.1 Reference Coils

In order to be able to tell if the array of six receive coils worked in a proper way, a reference coil had to be built. In this case, an array of two coils was used as a reference. This coil includes a decoupling circuit which is exactly the same as in the array of six coils. Because these two coils are much bigger than the six coils used before, higher values of C and L were used to tune and match the coil. See figure 29. The Q-spoiling circuits were built by using an inductor with L = 17.5 nH, a capacitor with C = 10 pF and a PIN-diode. The tuning capacitors have values between 2.8 and 12.5 pF. Capacitors with values between 9.8 and 60 pF were used to match the coils.

8.2 Imaging



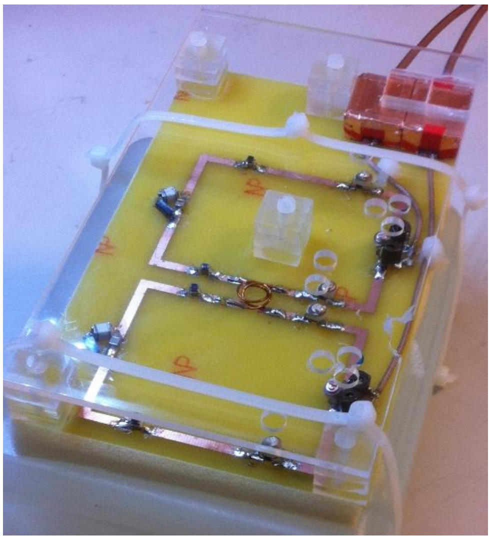


Figure 29: Reference coils used to compare with the array of 6 receive coils

8.2.1 Phantom Imaging

To compare the working of the array of six and the array of two coils, a phantom bottle was used. This bottle contained water and an amount of sodium chloride. In this case the concentration of sodium chloride in the bottle was 70 mMol/L. As was mentioned earlier, a 7 Tesla MRI scanner was used to do the imaging. By using this scanner, low tip angle gradient echo images were acquired [20]. At first, the array of two coils was used. The array was placed on top of the phantom bottle, as can be seen in figure 29. An example of an acquired image is shown in figure 30.

Figure 30: Low tip angle gradient echo image of a phantom bottle filled with a solution of sodium chloride with a concentration of 70 mMOl/L. An array of 2 receive coils was used to acquire the image.

After this, the array of six coils was attached to the phantom bottle in the same the array of two was in figure 29. The same imaging sequence was used and the result is shown in figure 31.

Figure 31: Low tip angle gradient echo image of a phantom bottle filled with a solution of sodium chloride with a concentration of 70 mMOl/L. An array of 6 receive coils was used to acquire the image.

8.2.2 In Vivo Imaging

To complete the imaging part of this research project, low tip angle gradient echo images of the lower leg were acquired. This was in vivo imaging by using the array of six receive coils. The participant was a 22 year old man. An example of an acquired image is shown in figure 32. The image clearly shows the two bones in the lower leg, the tibia and the fibula. There are also some visible blood vessels in the image, which are the white spots.

Figure 32: In vivo low tip angle gradient echo image of a lower leg. An array of 6 receive coils was used to acquire the image.

8.3 Interpretation & Conclusion

When looking at figure 30 and 31, it can be seen that the phantom bottle is brighter in figure 31 than it is in figure 30. This means two things. At first, it means that the signal picked up by the array of 6 coils is stronger than the signal picked up by the array of two coils. In other words, the signal to noise ratio (SNR) of the array of 6 coils is higher than the SNR of the array of two coils. The reason for this is the amount of coils used. So, more coils pick up more signal, as was predicted in the hypothesis. However, this increased signal is only useful if the decoupling of the coils is done in a proper way. In this research project, this was the case, as is shown in figure 26 in section 7.3. Secondly, the brighter signal does not only mean an increased amount of signal, but is also means that the coil can 'reach' deeper into the subject. In figure 31, the signal is not only brighter than in figure 30, it also reaches deeper into the phantom bottle. So it can be concluded that an array of six receive coils has a higher SNR and reaches deeper than an array consisting of two coils.

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