

Construction and Characterisation of MRI Coils for Vessel Wall Imaging at 7 Tesla



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A thesis submitted for the degree of Doctor of Philosophy

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Abstract

Atherosclerotic plaques in the bifurcation of the carotid artery vessels can pose a significant stroke risk from stenosis, thrombosis and emboli, or plaque rupture. However, the possibility of the latter depends on the structure of the plaque and its stability. So far, the assessment of such depositions, and the evaluation of the risk they pose, is not satisfactory with 3 Tesla black blood imaging. It is expected that the SNR increase at 7 Tesla, together with an appropriate and patient-safe RF coil, will result in higher resolution images that would help in better assessing the composition of atherosclerotic plaques in vessel walls. A custom-built neck array was designed and constructed, with the aim of investigating the benefits of the higher field strength using DANTE-prepared black blood imaging.

A 4-channel transmit array was designed to generate the required $\mathbf{B}_1^+$ field for the DANTE module to be used. A separate close fitting 4-channel receive array was preferred for improved SNR and parallel (receive) imaging. Geometric, active, passive as well as preamp decoupling schemes were employed for adequate isolation between the arrays and their channels. Electromagnetic simulation software, Semcad X (SPEAG, Zurich), was used for safety assessment with human phantoms (Virtual population). The $\mathbf{E}$ -fields for 1 W transmission per channel were calculated for each element for a worst case SAR estimation. The transmission power limits per channel were set according to the 10g SAR limit set in IEC 60601. For simulation validation, temperature measurements and surface heat mapping were performed on a meat phantom. Finally, a healthy male subject was scanned using a protocol consisting of $\mathbf{B}_1$ mapping, RF shimming at an ROI, and 2D and 3D DANTE-prepared Gradient Echo (GRE).

The worst-case heating scenario, as defined in the methods section, generated a maximum local SAR of 7.65 W/kg for 1 Watt per channel input. Thus, for 1st level mode (20W/kg max), the power limit was set at 2.6 W per channel. The heating profile was similar to that simulated and the measured temperature increase was within a $\pm 10\%$ margin relative to the simulation. The global SAR power limit per channel was found to be higher (i.e. more allowed power) than the worst case local SAR power limit, and thus did not impose additional power penalty. The resolution achieved was 0.6 mm isotropic for the 3D protocol and $0.6 \times 0.6 \times 2.5$ mm for the 2D protocol. The average SNR was measured within the vessel wall location of the two carotid arteries and found to be 27 ± 6 for the DANTE images and for the static tissue closer to the skin the SNR was 55 ± 2 .

In conclusion, a 4Tx/4Rx coil was designed to target the carotid arteries operating under pTx mode and a black blood imaging sequence was implemented for blood signal suppression and vessel wall imaging. The initial results from the subject and phantom imaging show satisfactory blood suppression and spatial resolution.

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Contents

1	Introduction	1
1.1	Motivation	2
1.2	Overview of the Thesis	4
2	Clinical Background	6
2.1	Cardiovascular System	6
2.2	Arterial disease	12
2.3	Vulnerable Plaque	16
2.3.1	Assessment of Vulnerable Plaque	17
2.4	MRI as a Diagnostic Tool	22
2.4.1	MRI in Atherosclerosis	22
2.4.2	MRI in Perfusion Imaging	26
2.5	Discussion	27
2.6	Summary	30
3	Principles of Magnetic Resonance Imaging	31
3.1	Introduction	31
3.2	Nuclear Magnetic Resonance	32
3.2.1	Larmor Precession	33
3.2.2	Free Induction Decay	34
3.2.3	Magnetisation Relaxation	36
3.2.4	Flip Angle	37
3.3	Magnetic Resonance Imaging	38
3.3.1	Slice Selection and Excitation	39
3.3.2	Image Contrast	42
3.4	Pulse Sequences and Image Formation	47
3.4.1	K -space	47
3.4.2	Spatial Encoding	49
3.4.3	Receiver Bandwidth	51
3.4.4	Pulse Sequence Trajectory	53
3.4.5	Functional Imaging	56
3.5	Instrumentation	60
3.5.1	The Magnet	60
3.5.2	Shim Coils	60
3.5.3	The Gradient Coil System	61
3.5.4	The Radio Frequency Coil	62

3.6	Development of the Radio Frequency Coil	65
3.6.1	Phased Array	67
3.6.2	The Birdcage Coil	67
3.6.3	Other Coil Concepts	69
3.7	Conclusion	70
4	Radio Frequency Coil Design Considerations	72
4.1	Introduction	72
4.2	RF Coil Design Theory	74
4.2.1	Tuning and Matching	74
4.2.2	Balancing the Probe	78
4.2.3	Parallel Transmission	82
4.3	MR Application	83
4.3.1	Black Blood Imaging - DANTE	83
4.3.2	Transmission Array Design	87
4.3.3	Receive Array Design	93
4.4	RF Coil Construction	100
4.4.1	Computer Aided Design	100
4.4.2	Decoupling Networks	101
4.4.3	Cable Preparation	114
4.5	Summary	120
5	Safety Considerations & Evaluation	121
5.1	Introduction	121
5.2	RF Safety	123
5.2.1	Specific Absorption Rate	124
5.2.2	Local SAR Limits	125
5.2.3	Other Hazards	133
5.3	System Description	134
5.4	4Tx/4Rx Coil Safety Validation	137
5.4.1	Model & Construction	137
5.4.2	Simulations & Averaging	138
5.4.3	Virtual Observation Points	142
5.4.4	Preparation and VOP clustering	145
5.4.5	VOP Validation	146
5.4.6	Global SAR	149
5.4.7	Worst Case SAR	155
5.4.8	Experimental Validation of the Coil	156
5.5	Other Safety Concerns	164
5.5.1	Common Mode Currents	164
5.5.2	Exposed Metallic Parts	165
5.6	Experimental Validation of Simulations	166
5.6.1	Methods	166
5.6.2	Results	174
5.6.3	Discussion	178
5.7	Conclusion	182

6	Coil Performance Evaluation	186
6.1	Simulation and Hardware Evaluation	186
6.1.1	Simulation Results	186
6.1.2	$\mathbf{B}_1$ -to-SAR ratio	187
6.1.3	Bench Measurements	192
6.1.4	Scanner Measurements	194
6.2	Flow Suppression Evaluation	197
6.2.1	Phantoms	197
6.2.2	Subject Scans	198
6.3	Summary	200
7	Discussion & Conclusions	216
7.1	Discussion	216
7.2	Future Work	218
7.3	Conclusions	220
	Bibliography	221

Chapter 1

Introduction

Blood flow restriction (ischaemia) in the brain due to atherosclerotic depositions in the carotid arteries is a subject of considerable clinical interest. Also, the rupturing of plaques in the carotid bifurcation with subsequent stroke is a complication that further motivates diagnosis and treatment options. As far as the diagnosis is concerned, arterial imaging using one, or a combination, of the clinical imaging modalities, is the most accepted method. The imaging protocol often comprises ultrasonography, X-ray computer tomography (CT) or use of magnetic resonance imaging (MRI).

Clinically approved MRI scanners are used for the assessment of the plaque vulnerability showing some clinical prospect [1, 2] that will be described in detail in the next chapter. However, the lack of high quality imaging at the approved magnetic strengths along with the technological improvements in MRI instrumentation has brought, over the past decade, the introduction of ultra high field MRI human scanners. Field strengths of double the clinically available field strength can generate higher quality images. The means by which this is possible will also be described in the second of the two introductory chapters in this thesis.

1.1 Motivation

MRI is used, among many other applications, in the diagnosis and assessment of atherosclerosis. Atherosclerosis is a potentially serious medical condition where the medium and large arteries become clogged up by fatty substances comprised mostly of cholesterol. The blood is pumped from the heart and through the aorta before travelling into smaller vessels that branch off to feed the vital organs. The hardening and narrowing of arteries can lead to restricted blood flow, reduced oxygenation and the death of cells in affected areas [3]. Furthermore, if a plaque ruptures (bursts) it may cause a blood clot or embolic mass that can travel via the circulation and block smaller vessels. That can cut off blood supply and trigger a myocardial infarction if the affected organ is the heart or a stroke if the blockage is located in the brain.

Stroke, in particular, is a very serious condition where the blood supply to a region of the brain is interrupted. Unlike the rest of the tissues, the brain cannot survive being starved of the oxygen and nutrients provided by blood for more than a minute or two. If the supply of blood is restricted or completely halted, brain cells begin to die within minutes causing irreversible brain damage and possibly death. According to UK Stroke Association, every year over 150,000 people in the UK have a stroke and it is the third largest cause of death, after heart disease and cancer.

In particular, the elementary lesion in the artery called atherosclerotic plaque consists of a lipid core (atheroma), a fibrous cap (sclerosis) and inflammatory cells. Histopathological studies have shown that plaques which caused symptoms were often ruptured, and had a thin fibrous cap and a bulky lipid core. The prevailing opinion is that this fibrous cap rupture is the factor that triggers thromboembolic complications. A possible trigger is the penetration of blood into the plaque at a small opening; such infiltration may subsequently give rise to a dissecting haematoma that will suddenly increase the plaque volume [4, 5].

Another consequence of such rupture is thrombosis, which is caused by blood contact with the plaque components which particularly contain the tissue growth

factor. The resulting thrombus size varies and can range from thin platelet size to a bulky mass and there are three options for its fate. One is migration (embolism), another is local growth which could lead to local occlusion and the third is self-healing. The first two possibilities can lead to severe complications and conditions as explained earlier with intra-plaque haemorrhage and very severe stenosis [2].

The solidity of the fibrous cap has a key role in determining the rupture risk. Soft plaques seem to be more unstable since they have a large lipid core that is only separated from the lumen by a thin fibrous layer. These plaques are often considered vulnerable posing a high rupture risk. The bifurcations of carotid arteries to the cerebral arteries are locations of high risk with respect to such ruptures since there is more shear stress at these locations. In addition, any separation of a clot will possibly cause a stroke. Thus, assessment of atherosclerotic plaque in carotid arteries is important so that appropriate treatment can be performed.

MRI is a non-ionising imaging technique and it is currently the most promising tool for non-invasive imaging of carotid vessel walls and assessment of atherosclerotic plaque. MRI is able to distinguish a lipid-rich core from a fibrous cap and thus to provide valuable information for the cap vulnerability. Additionally, the carotid artery is more accessible since it is less affected by physiological motion such as breathing and it is located close to the body surface [6].

All the MR vessel wall imaging studies have so far been conducted using clinically approved scanners that can have a maximum 3 Tesla field strength. Over the last decade higher than 3 Tesla research body scanners have been developed and acquired by various institutions in order to investigate the potential benefit of current and new imaging applications. The intrinsic MR signal is proportional to the field strength and thus the increase of the latter will lead to higher resolution images. However, it is a simplification due to the fact that there are repercussions from increasing the field strength, ranging from the spatially imbalanced images, i.e. $\mathbf{B}_1$ field inhomogeneity, as well as complications with RF safety.

The aim of this thesis is thus to systematically approach these problems for the application of vessel wall imaging, to propose solutions and to generate results that use the full potential of the new generation of scanners. The main aims of this thesis are, (i) to design and construct suitable and patient-safe RF coils at 7 Tesla; and (ii) perform high resolution imaging of the carotid artery at 7T.

Finally, the overall motivation of this thesis is to contribute in the development of a clinical modality that will allow for the accurate assessment and diagnosis of arterial plaques via high resolution Ultra High Field MRI (UHF-MRI), i.e. field strengths above 3T.

1.2 Overview of the Thesis

Following the preceding motivation and overall aims, an overview of the thesis will next be provided. Thus, in the next chapter, the relevant clinical background will be further described in detail and will shed light on the clinical need and the motivation behind this study. In Chapter 3 a short but necessary theoretical scientific background is provided, i.e. basic principles of MRI including a section on the radio frequency system, its function and historical development.

The main content of Chapter 4 is the design considerations for the RF coil as well as its construction. The selected application will be analysed with regard to the RF field requirement along with other factors such as the average anatomical location of the carotid arteries and the blood velocity, and from this a design will be proposed. Electromagnetic simulations are an important tool in RF design and they are extensively used in this chapter. Finally, Chapter 4 includes a technical description of the construction processes that were followed to build the hardware. The chapter details the electrical components, circuit diagrams, decoupling schemes and RF interference elimination techniques.

Chapter 5 refers extensively to RF safety during scans. The most challenging

problem that this study faced was the approval of surface transmission arrays for human scanning. The problem is described systematically and was approached conservatively with the worst case SAR approximation. A more accurate approach is also investigated, the Virtual Observation Points (VOPs), where local power deposition is monitored during the scan. The last section of the chapter includes an important process in safety evaluation which is the simulation validation with temperature measurement experiments.

In chapter 6, the coil is evaluated as an RF device as well as an imaging coil. SNR maps, $\mathbf{B}_1^+$ and $\mathbf{B}_1^-$ maps, g-factor maps, noise correlation matrices, bench measurements, such as scattering parameters for each mode and for every loop combination, are all included and discussed. Healthy consenting and suitable volunteers were scanned and the performance of the coil for the proposed applications was assessed. Finally, a general discussion, future work section along with the conclusions are included in Chapter 7.

Chapter 2

Clinical Background

In this chapter, the relevant anatomical and physiological details are outlined from the literature to help the reader understand the background of the clinical problem and the validity of the proposed solutions. Also, the general purpose and motivation of the study are discussed.

2.1 Cardiovascular System

The main element of the cardiovascular (or circulatory) system is the heart, which contracts and relaxes rhythmically its four compartments (two atria and two ventricles) as well as its four valves in order to pump blood around the body. The vasculature is divided into two groups, the pulmonary and systemic circulations. The former is embedded within the lungs and its purpose is to enrich the red blood cells with oxygen and to return them to the heart before it is pumped around the systemic circulation. The arteries are the blood vessels that carry the blood away from the heart and the veins are the vessels that carry the blood towards the heart. In the systemic circulation the arteries carry oxygen-rich blood and the veins carry de-oxygenated blood including metabolic waste. In the pulmonary circulation the latter is reversed. An illustration of the circulation is presented in figure 2.1. A more thorough description can be found in [7].

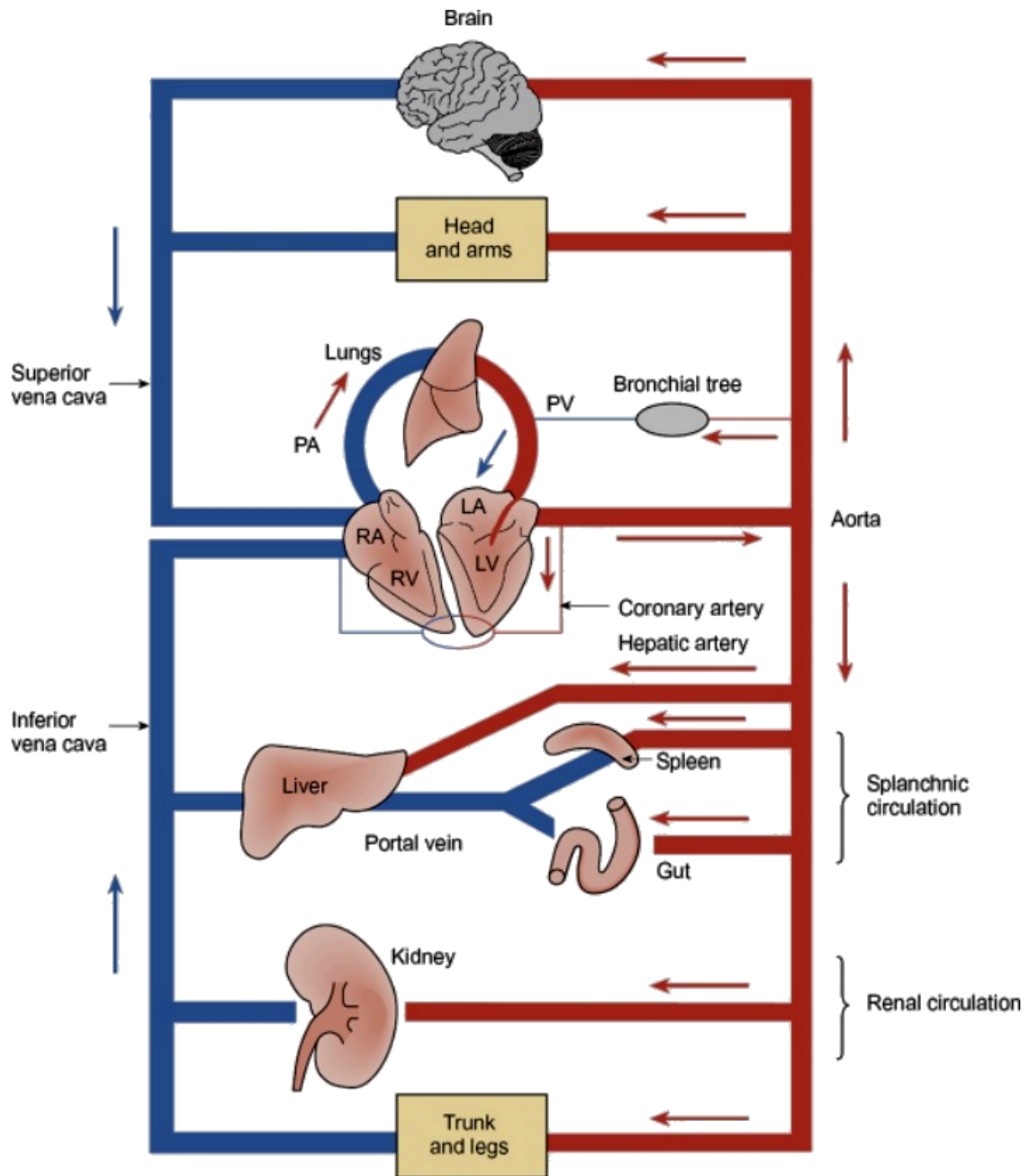


Figure 2.1: Cardiovascular system. Systemic (cerebral, lower body) and pulmonary circulation systems are shown. Reproduced with permission from [8].

The other main elements of the circulatory system are blood and vessels. Blood is mainly formed of plasma and blood cells, which can be divided into several subgroups with the most significant being red blood cells and white blood cells. An important function of the white blood cells is to fight off infections in the human body and to initiate the healing of damaged tissue by forming local blood clots and inflammation. This body function can have detrimental side effects when it occurs in vessels with no pathogenic trigger and they will be discussed in detail later.

As mentioned, large vessels are divided into arteries and veins and they have different anatomical properties due to their intrinsic function. In particular, veins carry blood towards the heart, they are less muscular, less flexible than arteries and have directional flow valves. Arteries carry blood away from the heart and their walls are more responsive to blood pressure and body function via vasodilation or vasoconstriction. The vessel layers, starting from the outermost, are called adventitia, external elastic lamina, smooth muscle, basement membrane, internal elastic lamina, endothelium and lumen. The structure and thickness of each layer varies depending on the location in the body for any vessel. A side by side illustration of the two vessel types can be found in figure 2.2.

The endothelium is the innermost thin layer of cells in blood vessels. The endothelial cells are in contact with blood and perform vital functions of the vascular system. A few of these functions are fluid filtration, haemostasis and hormone transportation. The compromise or complete loss of any of these functions is associated, amongst others, with an early event in the development of atherosclerosis, a disease responsible for many deaths worldwide [9]. More information about endothelial function and its association with disease can be found in [10, 11].

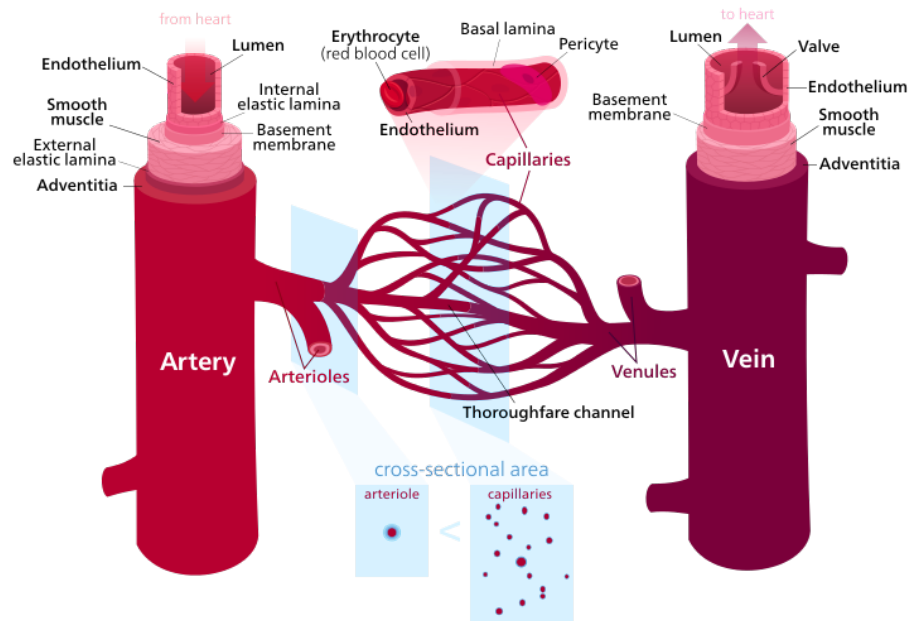


Figure 2.2: Artery and vein layers, their connection through the arterioles, venules, capillaries. Reproduced from [12].

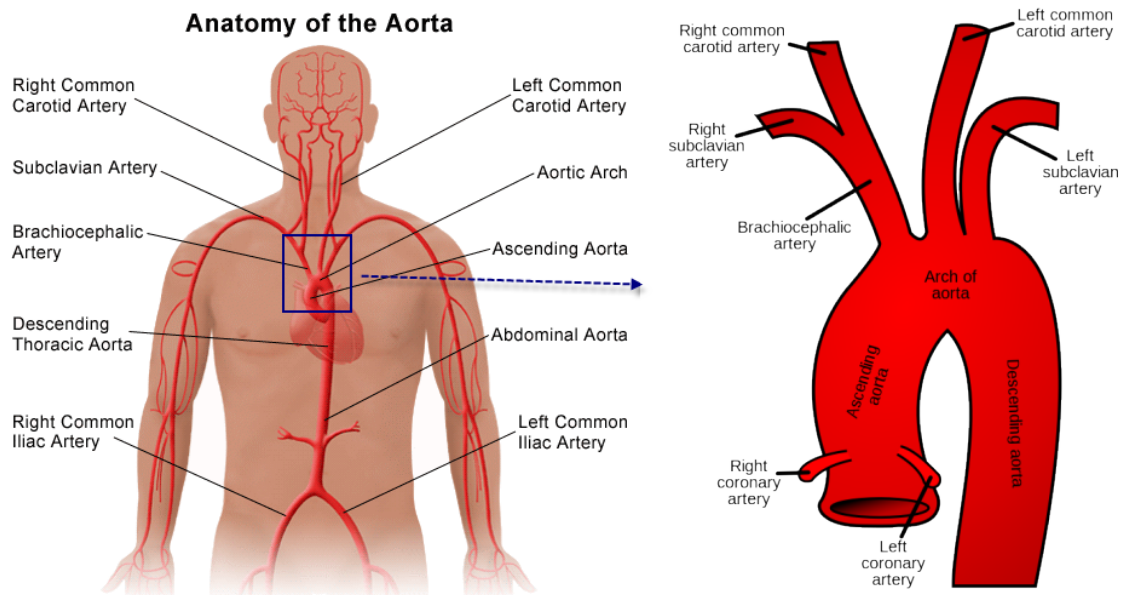


Figure 2.3: Ascending and descending aorta and branches representation. Reproduced with permission from [13].

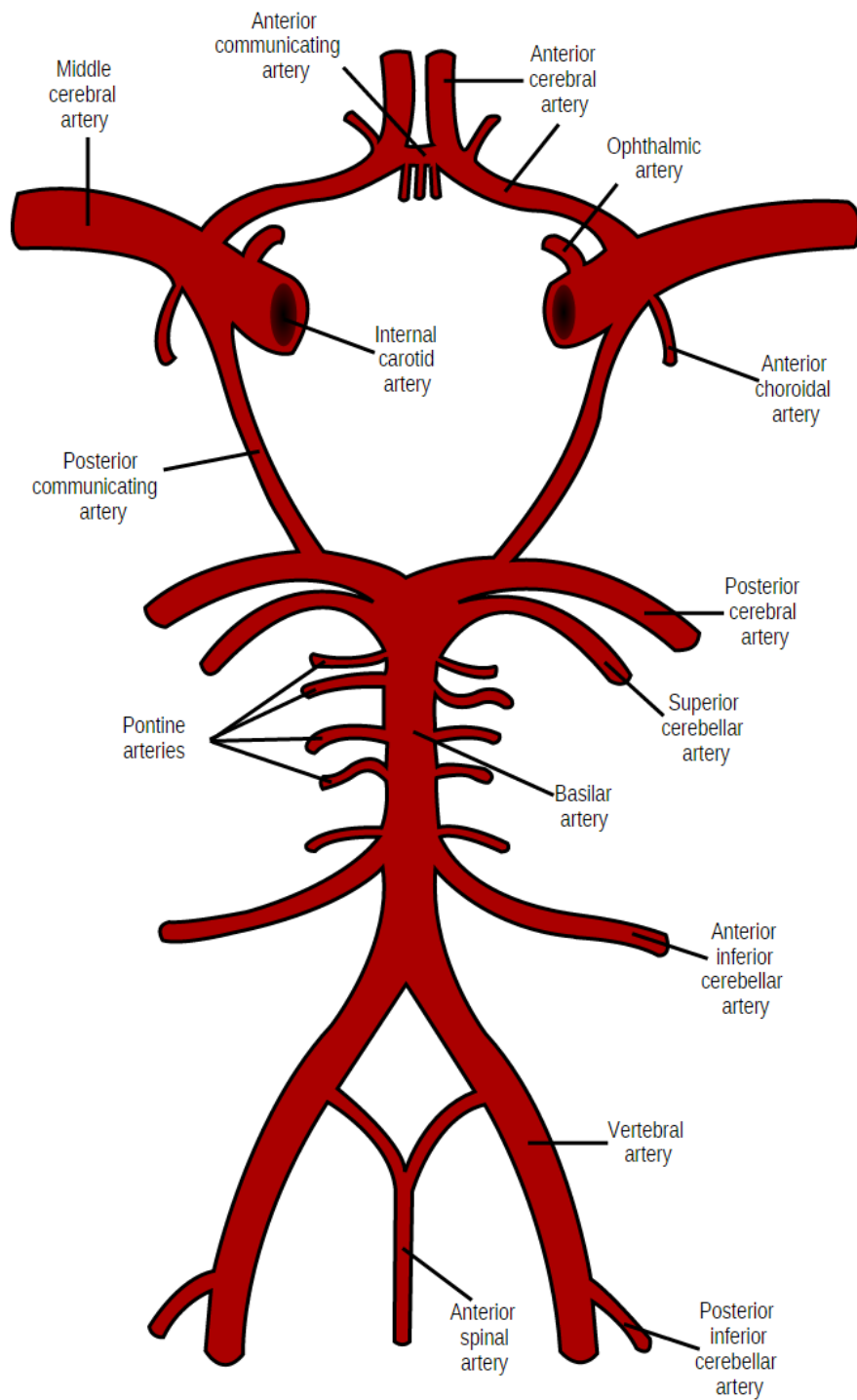


Figure 2.4: Detailed representation of part of the cerebral vasculature. Reproduced from [14] with permission.

The left and right common carotid arteries (CCAs) are the arteries that carry oxygen-rich blood to the brain. In the neck area and just before the blood arrives in the cranial area the CCAs bifurcate to two smaller arteries, the internal and the external carotid arteries, known as the ICA and the ECA respectively. The exact location of the bifurcation in the neck region varies among subjects in skin depth and relative height following a normal distribution around the C4 vertebral disc with C2 and C5 to be the 1% of the population as illustrated in figure 2.5. This study focusses on carotid artery disease and in particular critical thromboembolic events that originate from atherosclerotic plaques in the vicinity of the bifurcations. A detailed outline of arterial disease is thus given in the next section.

In this study our focus is on the cerebral system i.e. the vasculature that perfuses the brain. Blood flows in to the brain via the neck but it originates from branches in the largest vessel in human body, the main aorta. The aorta comes out of the top of the heart, hence being called the ascending aorta, making a U-turn (known as the aortic arch) to direct blood down to the lower body. On the top of the aortic arch there are three arterial branches, the brachiocephalic artery, the left common carotid artery (L-CCA) and the left subclavian artery. The brachiocephalic artery then branches into the right subclavian artery and the right common carotid artery (R-CCA). An illustration of this vascular anatomy can be found in figures 2.3 and 2.4.

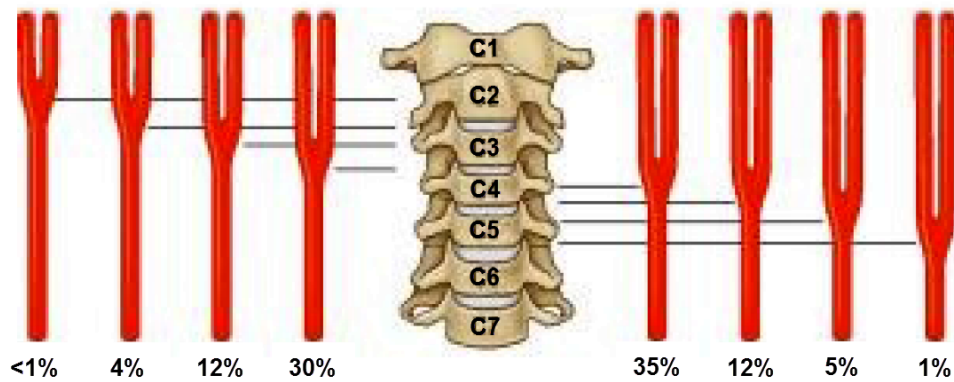


Figure 2.5: Variation in levels of bifurcation of the common carotid artery, related to the cervical vertebrae. Reproduced from [7].

2.2 Arterial disease

The cardiovascular system can be affected by various conditions or diseases and in turn can gravely affect vital organs in different ways. The most common and extensively studied disease of the arteries is atherosclerosis. This is the thickening of the vessel walls due to white cell invasion and the creation of inflammation. Due to local inflammation there is entrapment and accumulation of cholesterol, triglyceride molecules, calcium and other blood components, which eventually form a solid wall thickening which after time affects the elasticity and function of the vessels [15, 16].

The onset of the disease is not fully understood but it is thought to be due to the entrapment of lipoprotein particles below the endothelium wall and subsequent oxidation [17]. In particular, its occurrence has been linked to the blood levels of low density lipoprotein (LDL) vesicles. Lipoproteins enable the transportation of lipids such as cholesterol and triglycerides within the aquatic environment of the extracellular fluid, particularly the blood plasma. There are various types of lipoproteins with different diameters but the LDL diameter happens to be in the range of the endothelial cell opening, enabling it to pass through the endothelial monolayer. However, there is a possibility that it cannot find an opening wide enough to be freed and it can thus be retained behind the endothelium. The High

Density Lipoproteins (HDL) on the other hand are smaller than the openings and can pass freely in and out and on occasions it can help to clear the stuck LDL particles off the wall. This is the reason why the LDL is sometimes considered ‘bad’ cholesterol and HDL ‘good’. A summary of the average densities in the blood levels, diameters and compositions can be found in the table in figure 2.6.

Density (g/mL)	Class	Diameter (nm)	% protein	% cholesterol	% phospholipid	% triacylglycerol & cholesterol ester
>1.063	HDL	5–15	33	30	29	4
1.019–1.063	LDL	18–28	25	50	21	8
1.006–1.019	IDL	25–50	18	29	22	31
0.95–1.006	VLDL	30–80	10	22	18	50
<0.95	Chylomicrons	100-1000	<2	8	7	84

Figure 2.6: Classification of the different types of lipoproteins. The LDL molecule has a diameter of 18-28 nm posing a risk for adherence behind the endothelium. On the other hand, the HDL has a smaller diameter and is thought to help clear away any trapped LDL molecules from the endothelium as well as fats and the build-up of macrophages hence decelerating the build-up of plaque [18].

Adherence inside the endothelium does not necessarily cause disease. However, lipoprotein particles and their content, including LDL, are susceptible to oxidation by free radicals and the risk is higher while they are in the endothelium wall [19]. Endothelial cells respond by attracting monocyte white blood cells, causing them to leave the blood stream, to penetrate into the arterial walls and to transform into macrophages. The macrophages’ ingestion of oxidized LDL particles triggers a cascade of immune responses which over time can produce an atheroma. The formation of the latter depends on the HDL fat removal rate and how it compensates for the plaque build up.

The immune system’s specialized white blood cells (macrophages and T-lymphocytes) absorb the oxidised LDL, forming specialised foam cells. If these foam cells are not able to process the oxidized LDL and to recruit HDL particles to remove the fats, they grow and eventually rupture, leaving behind cellular membrane remnants, oxidized materials, and fats (including cholesterol) in the artery wall. The latter

attracts more white blood cells, resulting in an avalanche progression that continues to inflate the artery.

The additional bulk from the plaque causes endothelial lining thickening, increasing the separation between the plaque and lumen and forcing the muscle cells of the blood vessel to stretch. This mechanism compensates for the narrowing caused by the growth of the plaque, but it causes the wall to stiffen and to become less compliant to stretching with each heart beat. The disease can be asymptomatic for decades and the onset of the initial lesion can be as early as childhood. Figure 2.7 shows how atherosclerosis evolves, the earliest onset and the microstructure of the plaque at each stage.

Atherosclerosis thus hampers the function of the arteries causing them to lose their flexibility. The initial symptoms are associated with loss of stamina and shortness of breath, because the arteries are unable to dilate and to allow more blood to circulate. The next stage as the plaque grows is critical stenosis that can restrict the blood flow causing ischaemia. If the arteries affected are the ones supplying blood to the heart, coronary arteries, the cardiac muscle will deteriorate and possibly muscle cells will die over time leading probably to an irregular heart beat rhythm and loss of sustainable blood flow in the body (myocardial infarction).

One other detrimental effect that can be caused from a growing plaque is a partial break down of the lesion and release of particles into the blood stream. These can travel in the system until they reach a narrow vessel and block it causing complete or significant partial ischaemia to the tissue fed by the particular vessel. This thrombosis is a serious complication that requires immediate treatment. If the clot enters the brain then it can cause a stroke, a life-threatening condition with unpredicted damage in brain function. In the next section, the vulnerable plaque will be thus discussed in detail.

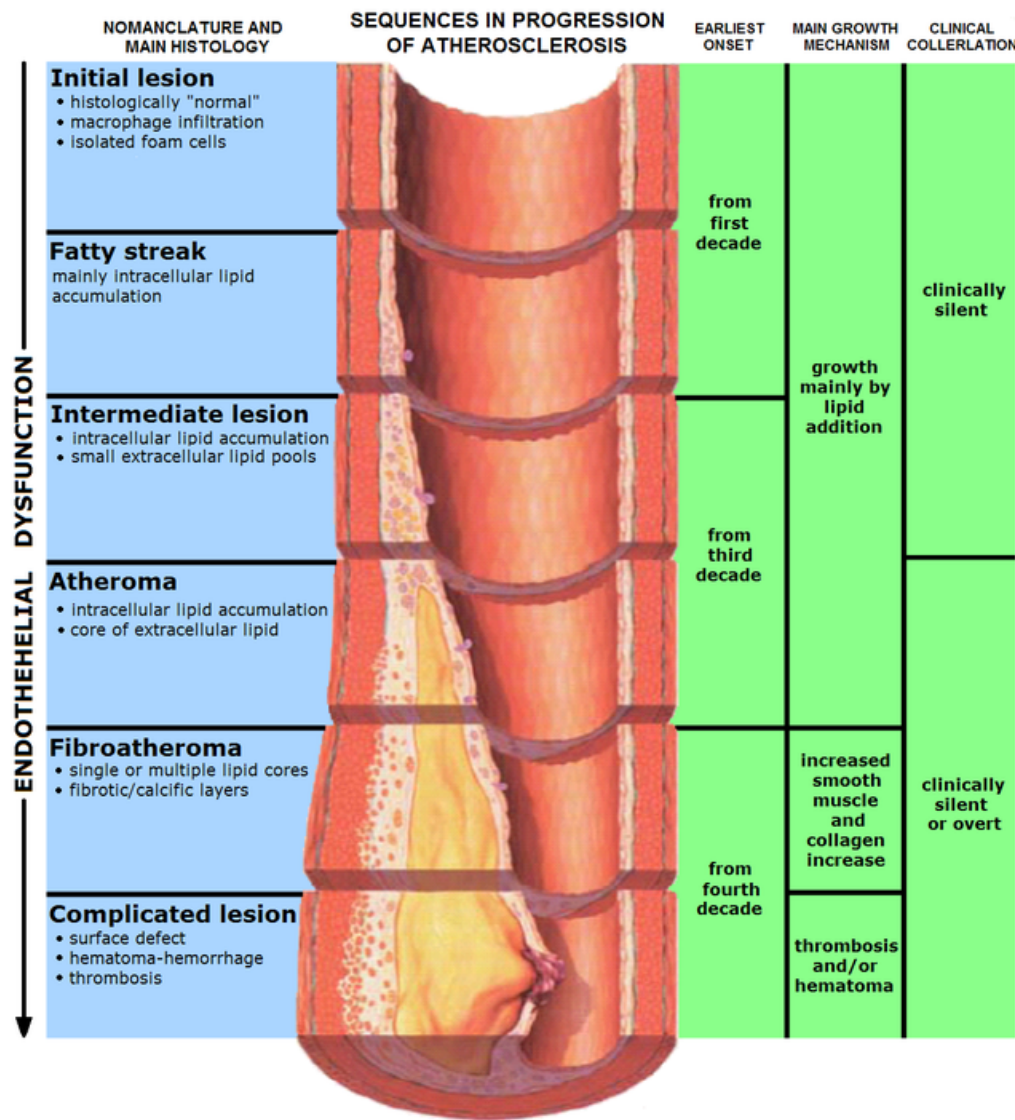


Figure 2.7: Progression of atherosclerosis. It can take up to two decades of clinically silent atheromatic build-up with the final stages causing blood flow restriction, lesion and possible thromboembolic events. Image reproduced with permission from [20].

2.3 Vulnerable Plaque

There is extensive debate regarding the causes and risk factors of atherosclerosis but this study concentrates on the diagnosis and assessment of the plaque stability in the advanced stages of the disease. In figure 2.7 it is clear that as the plaque enlarges over time, its inner composition changes. In particular, when it progresses from lesion to atheroma and further, a distinct core of extracellular lipid is formed. What separates the core from the blood in the vessel is a layer called the fibrous cap that has a thickness that can vary from person to person.

The fibrous cap is a layer of connective tissue containing macrophages, bundles of muscle cells, foam cells, lymphocytes and elastin [21]. Endothelial activation and infiltration of monocytes transforming to macrophages, i.e. repair cells, instigate the migration of smooth muscle cells and the formation of the fibrous cap. In some cases the macrophages are unable to cope with the increased concentration of lipids and die, releasing their contents, which can then attract more macrophages. Increased activity of these cells can induce erosion of the fibrous cap due to increased enzyme presence associated with macrophages, thus thinning the fibrous cap. The mechanical stretching from the heart beat and the blood pressure can bring the structural integrity of the layer to a critical condition, towards rupturing and releasing of the contents of the atheroma with serious repercussions [22].

In figure 2.8 the different types of vulnerable plaque as the underlying cause of thromboembolic events are illustrated. The composition of each of the parts constituting the plaque determines its stability. The plaque component proportions can vary among different plaques. Rupture-prone plaques in the coronary arteries usually have a thin fibrous cap (cap thickness $< 65\mu m$), a large lipid core ($> 40\%$ of the total lesion), and high degree of inflammation [23, 24]. However, critical plaques in the carotid artery are characterised by severe stenosis and are also mainly fibrotic [25]. It is understood that the composition of a lesion can depend on its anatomical location, with significant variation among different or even the same patient. It is

thus clinically desirable to use or to develop reliable non-invasive imaging methods in order to diagnose and characterize atherothrombotic vulnerable plaques for each anatomical region, and with the ability to date their progression stages [1].

2.3.1 Assessment of Vulnerable Plaque

Many diagnostic modalities are used for the diagnosis and assessment of atherosclerosis. As explained, the disease can be asymptomatic for decades and hence it is unlikely to be diagnosed in the early stages. However, populations that are likely to develop the disease are usually subscribed to screening programs that start from simple home based questionnaires to intravascular ultrasound or other interventional examinations. The pyramid of vulnerable patients, figure 2.9, shows the type of screening and the corresponding number of patients screened in the United States.

Angiography is a generic term for imaging the vessels of the body and is a common medical imaging method for the visualisation of the blood vessels. It is an interventional procedure since it requires the injection of X-ray absorbent solution (iodine based contrast material) to the circulation. The imaging is done via 2D X-rays, 3D imaging with CT scanning or even 4D imaging, i.e. including the time dimension. The last two methods are commonly referred to as CT angiography. The angiograph, depending on the anatomical location and the resolution requirements, can be invasive, such as coronary catheterisation that entails the insertion of a catheter through the femoral artery for the local administration of the contrast agent. In contrast to conventional angiography, CT angiography visualises the vessel wall as well as the plaque composition, see figure 2.11. Thus, CT scans have produced valuable results so far in regards to the assessment of the vulnerable plaque [26]. When compared to clinically approved MRI, CT resolution is higher.

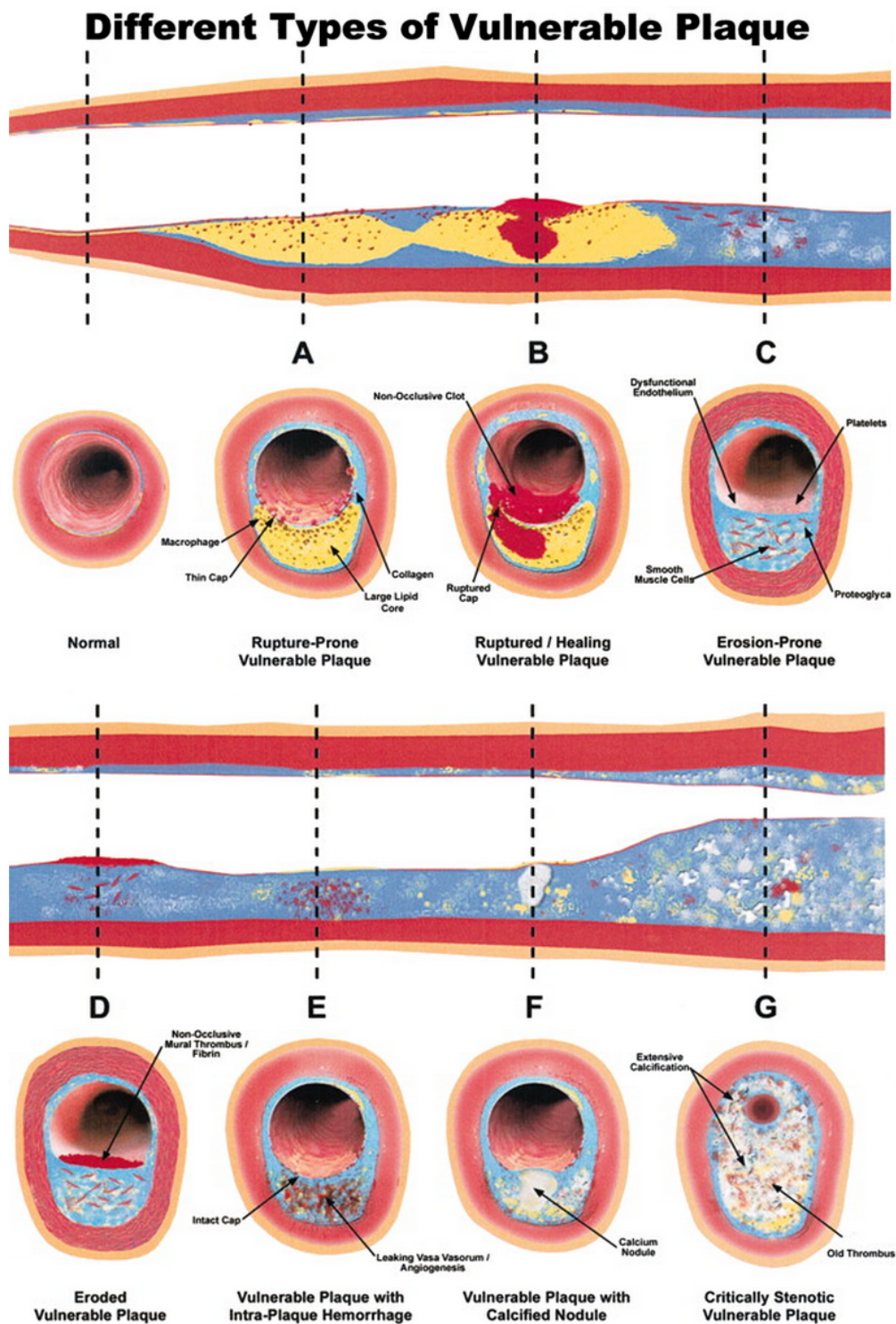


Figure 2.8: Different types of vulnerable plaque. A: The main observation on a rupture prone plaque is the very thin fibrous cap (in blue) and a large lipid core (in yellow). B: Healing plaque after rupture, blood clot forming in and on top of the fibrous cap. Clot is prone to separation and causing thrombosis. C: Different type of plaque core, smooth muscle cell-rich, erosion-prone plaque that can lead to a haematoma as observed in D. E: Intra-plaque haemorrhage in the core, F: Calcific nodule protruding into the vessel lumen. G: Stenotic plaque with severe calcification, old thrombus, and eccentric lumen. Reproduced with permission from [4].

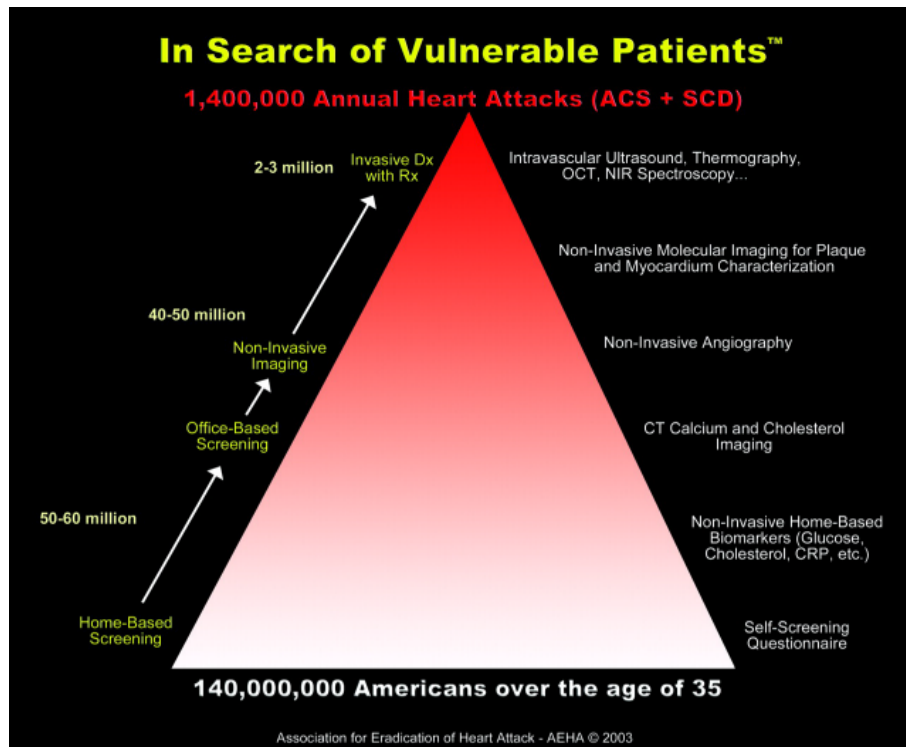


Figure 2.9: The Vulnerable Patient Pyramid illustrates a risk evaluation of patients with atherosclerosis in the United States. The bottom of the pyramid includes all adults over 35 years old using plain diagnostic tools allowing increasingly invasive and thorough screening of fewer patients with atherosclerosis and a risk for acute events. Each diagnostic method mentioned here will be discussed later. Reproduced with permission from [5].

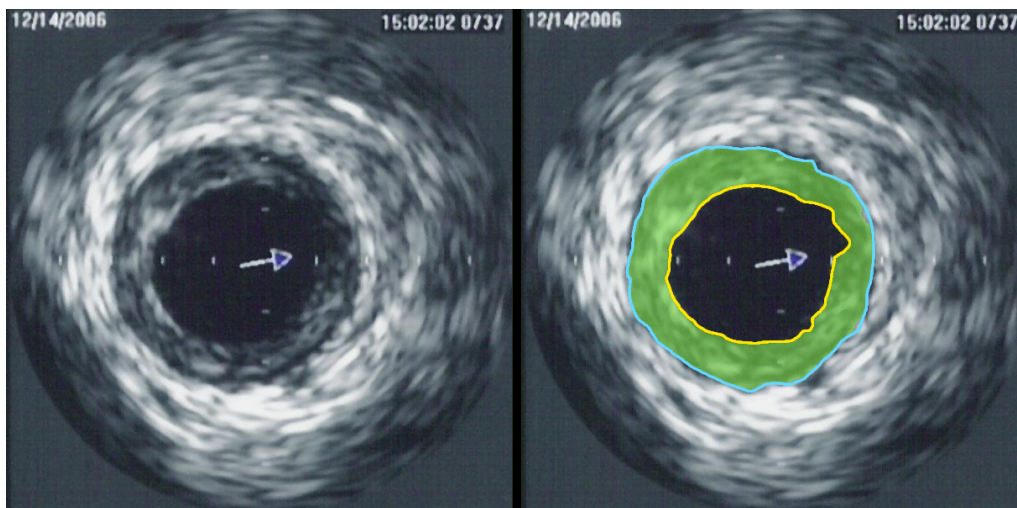


Figure 2.10: Intravascular Ultrasound (IVUS) image of a coronary artery (left), with colour coding on the right. The lumen boundary is depicted in yellow, the external elastic membrane in blue and the atherosclerotic plaque in green. The stenosis fraction is defined as the area surrounded by the lumen over the area of the external elastic membrane and usually given as a percentage. Reproduced from [27].

Since artery walls typically enlarge in response to enlarging plaques, there is not much evidence of stenosis just by imaging the artery lumen. Therefore, they are not reliably detected by classic angiography but only via CT. However good the imaging results, the nature of the contrast agent used poses a risk for allergic reaction as well as excluding patients with kidney disease or diabetes for a CT angiogram. Also the exposure to ionising radiation is higher than a standard CT scan, bringing the dose to a level that can increase the possibility of cancer occurrence within the lifespan of the patient [28].

Intravascular ultrasound is another imaging modality used for the assessment of plaque stability. Minimally invasive, an ultrasound angiography catheter is inserted in an artery or vein and the active tip captures the lumen of a specific location in the vein. Arguably the most valuable use of intravascular ultrasound, or IVUS, is to visualize the plaque, which cannot be seen by plain angiography (only by CT). It has been increasingly used in research to better understand the behaviour of the atherosclerosis process in living people [29, 30, 31].

However, IVUS is also used as a follow-up of an angiographic examination, such as a CT scan, when it gives ambiguous results, indicating that some locations need further assessment, hence being a modality to deploy in combination with angiography and not alone, see figure 2.10 for an example. It is a relatively expensive method and lasts much longer than the typical catheter assisted angiography, entailing for that reason an additional risk to the patient. Also, highly specialised medical personnel to perform the procedure are not available in many locations.

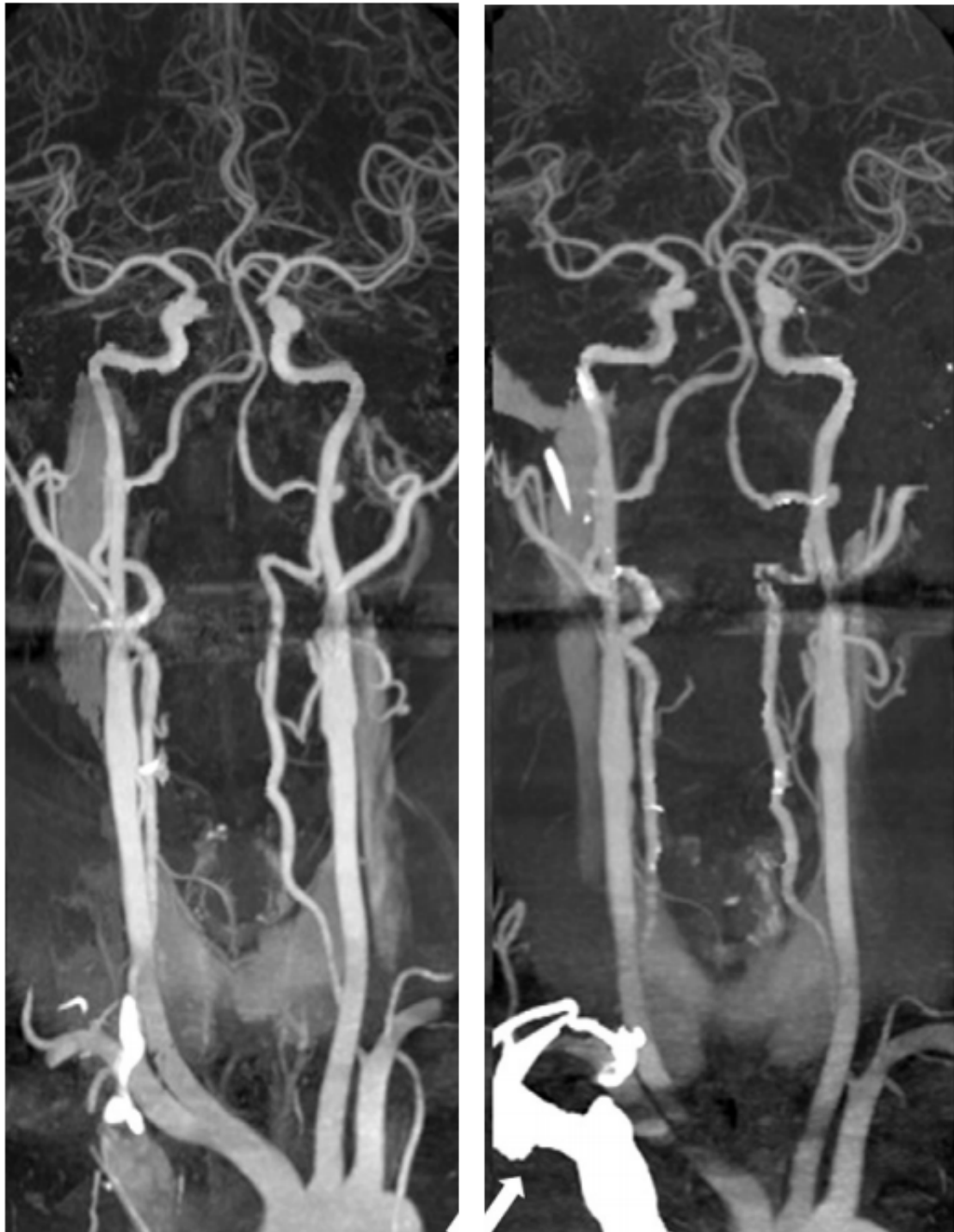


Figure 2.11: Maximum Intensity Projection (MIP) images of CT angiography of the carotid artery. The reconstruction covers the area from the aortic arch to the top of the cranium. The carotid arteries as well as the circle of Willis can be clearly seen. Reproduced with permission by [32].

Another approach to detecting and understanding plaque behaviour, used in research and by some clinicians, is to use conventional ultrasound to non-invasively measure vessel walls or intima-media thickness (IMT) in portions of larger arteries closest to the skin, such as the carotid or femoral arteries. More information can be found in the following studies [33, 34, 35]. Other methods for the imaging of the blood vessel walls include thermography and near-infrared spectroscopy to predict these lesions and the individuals most prone to atherosclerosis related episodes but they have remained only at a research level with no useful clinical results as yet.

2.4 MRI as a Diagnostic Tool

2.4.1 MRI in Atherosclerosis

In the absence of an adequate non-invasive tool for imaging atherosclerosis, the chance of diagnosis during screening is reduced, leading to late prevention action or insufficient treatment. Ideally, an imaging method should be safe, i.e. non-invasive and not involving ionising radiation or contrast agents, inexpensive, accurate and reproducible. Thus, high resolution magnetic resonance imaging (MRI) has emerged as the most promising modality for the diagnosis of atherothrombotic disease. MRI allows for the characterisation of the plaque composition, i.e. the distinction between the lipid core and the fibrous cap as well as the composition of these compartments. Magnetic resonance angiography provides information on the location and severity of stenotic lesions and localised high-resolution imaging provides information on the plaque composition, see figure 2.12.

The reliability of MRI techniques in atherosclerosis diagnosis has gradually increased through multiple pre-clinical studies validating initially the accuracy of the geometrical findings and later identifying the plaque composition with respect to the contrast provided in MRI. In [36] patients scheduled for endarterectomy, i.e. the removal of arterial sections with stenotic plaque, were MRI scanned. The carotid

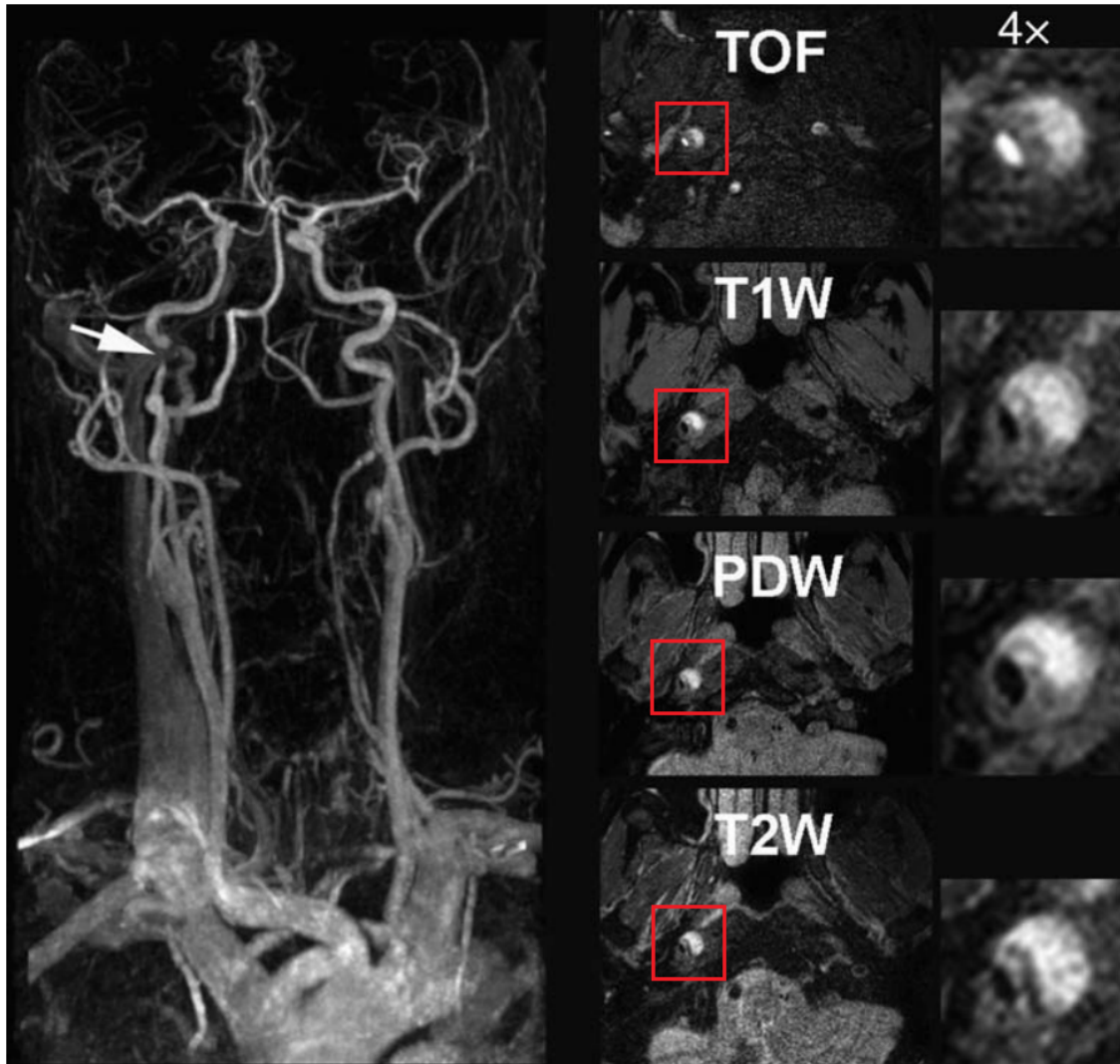


Figure 2.12: Magnetic resonance angiography demonstrating a stenosis finding of the right internal carotid artery (arrow). On the right column four different MR imaging techniques are used with contrast mechanisms that will be discussed in the next chapter. Reproduced with permission from [1].

artery wall dimensions as seen on MRI were compared to the excised arterial tissue. The comparison of the two measurements strongly agreed, validating the geometrical accuracy of the high resolution MRI for vessel wall imaging. This study showed that with MRI it is possible to monitor lesion and plaque size and progression over time.

The first experiments in-vivo for the imaging of plaque were performed in small animals such as rabbits and mice [37, 38, 39]. The lesions in the animals were dissected and their structure compared to the previously acquired MR images. The plaque fine structure was verified with histology and microscopy was assessed as to how it correlates to the image. The contrast achieved for the different layers of the plaque, as described previously, was deemed satisfactory showing that the MR signal can differentiate the components in the plaque. The study using a rabbit as the test subject had similar findings. For assessment of the progression of the lesion, 1.5 Tesla scans were acquired 9 months after the diagnosis of the initial aortic atherosclerotic lesion and again there was found to be a match with the histopathological findings. These studies undoubtedly have shown that MRI is a reliable tool in diagnosing and assessing atherosclerosis over time.

The transition from pre-clinical models to human subject studies was done through the use and comparison of dissected lesions after endarterectomy of patients with diagnosed critical atherosclerosis. In other studies it was demonstrated that specific components in atherosclerotic plaques can be distinguished with MRI [40, 41]. Furthermore, Toussaint et al. [42, 43, 44] concentrated on MR imaging techniques for the identification of the fibrous cap characteristics, concluding that calcification, fibrous intimal tissue and intra-plaque haemorrhage can be identified with in-vivo T_2 contrast measurements of carotid arteries. The available MR contrasts will be discussed in more detail in the next chapter.

Carotid arteries in particular are suitable for MRI due to their superficial location, adjacent to the skin (approximately 3 cm deep) and limited disturbance from

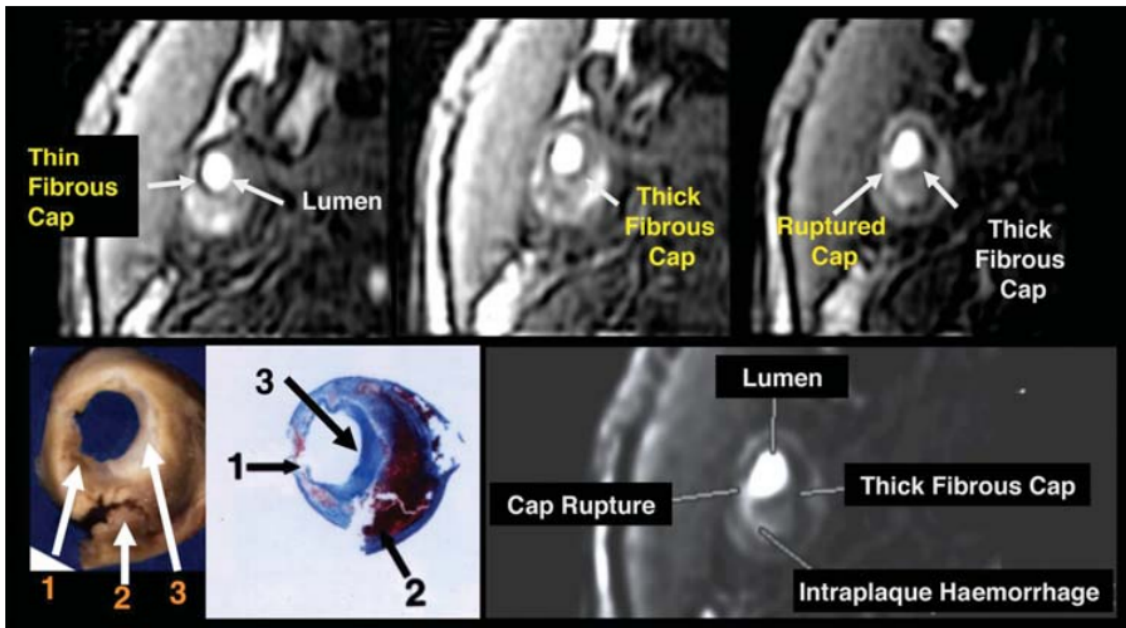


Figure 2.13: Example of MR angiography and histological sections of a carotid artery section with atherosclerotic plaque taken from the same subject after carotid endarterectomy. Arrow 1 shows the stenosis, the intraplaque haemorrhage pointed by arrow 2 and the thick fibrous cap, arrow 3. Reproduced from [45, 48, 1] with permissions.

motion. Some of the carotid plaque MRI studies have attempted both imaging and characterization of normal and pathological arterial walls, the quantification of plaque size, and the detection of fibrous cap integrity [36, 45]. In figure 2.13, images taken in a 1.5T scanner show the section of the atheromatic artery.

Although not yet available for routine use, in vivo, high-resolution, multicontrast MRI remains the most promising method of non-invasively imaging plaques and characterizing the main plaque components. Intraplaque haemorrhage and the lipid-rich necrotic core are the best indicators of lesion severity currently visualized by high-resolution MRI. However, MRI methods capable of imaging other important aspects of atherosclerotic disease in vivo, including inflammation, neovascularization, and mechanical forces, are emerging and may aid in advancing our understanding of the atherothrombotic disease [46, 47].

2.4.2 MRI in Perfusion Imaging

Perfusion is the estimation of how much blood is delivered to the tissues in any given time. It is an important vital sign and low perfusion is an early indication of an ischaemic event. In MRI, perfusion imaging is possible and it refers to the capillary blood supply to a tissue, measured in $mlmin^{-1}g^{-1}$ and in brain it is called the cerebral blood flow (CBF) or f . Two other measures are commonly used in perfusion MRI, the cerebral blood volume (CBV) and the mean transit time (MTT). These terms are usually prefixed with ‘r’ (rCBF, rCBV and rMTT) meaning ‘relative’, since their absolute quantification is difficult and hence ratios between the brain sides or compartments are often used instead.

Two MRI techniques for imaging perfusion have been developed in the last few years, with or without using a contrast agent to distinguish perfused tissue from unperfused, primarily for neurological applications. The quicker method is to image the required slices rapidly and repeatedly during the first pass of a gadolinium bolus, sometimes referred to as ‘dynamic susceptibility contrast MRI’ (DSC-MRI). The other technique tags protons in the arterial blood supply with a magnetic ‘label’, then images of the required slices with and without the labelling are subtracted. Although a slower technique with poor signal-to-noise ratio (SNR), it is completely non-invasive and can be repeated as often as required (without having to wait for excretion of the gadolinium). It is known as ‘arterial spin labelling’.

In DSC-MRI a volume of tissue, usually only a few slices, is imaged repeatedly and after a few images have been collected as a baseline, a dose of gadolinium is injected. During the first pass through the intracranial circulation the gadolinium remains in the vasculature, causing a reduction of the relaxation constants of blood, T_2 and T_2^* , which results in a drop in signal intensity in the image. The second pass may also be detected as a slight drop in intensity, before the signal returns to baseline. The whole imaging sequence takes a few minutes. The contrast agent injection should be started soon after the start of the imaging sequence which is an added

complication in the process. Full analysis for quantification uses deconvolution of the shape of the injected bolus, known as the arterial input function (AIF). This can be measured in the middle cerebral artery or any other easily identifiable vessel, but caution should be exercised in patients with stroke who may not have a normal arterial supply. For more information the reader can refer to [49, 50, 51].

Arterial Spin Labelling (ASL) imaging is based on magnetically labelling blood water protons in an appropriate arterial location that supplies blood to the region of interest, e.g. carotid arteries for brain perfusion. Images are then acquired in the RoI with a subtle signal difference compared with unlabelled images. Due to relaxation decay fast imaging is required which entails low SNR and thus low spatial resolution which is the most prominent disadvantage of ASL. Hence higher field strength imaging could lead to a better ASL performance additionally benefiting from longer T_1 constants. There are several different ASL techniques, differing mainly in the way they apply the labelling and control pulses. A short discussion in Arterial Spin Labelling will be made in the next chapter but for more information the reader can refer to [52, 53, 54, 55, 56, 57, 58, 59, 60, 61].

2.5 Discussion

The availability of robust non-invasive techniques to image atherosclerotic vessels has the potential to allow comprehensive study of atherosclerosis distribution and plaque risk to the cerebrovascular system. The limitation of non-invasive imaging the vessel lumen by MRI versus CT is mainly the spatial resolution that is lower, based on clinically approved scanners, than the plaque fine structure spatial variation observed under the microscope.

Spatial resolution in radiology refers to the ability of the imaging modality to differentiate two objects. Low spatial resolution techniques will be unable to differentiate between two objects that are relatively close together. In digital imaging

tomography the resolution matrix or grid is chosen by the investigator and each pixel comes with one value or signal intensity. The best matrix choice would approximate the ‘real’ resolution as defined above and in a medical imaging modality is dictated by the intrinsic signal strength. There are numerous parameters that can be optimised to extract the best possible resolution in the most feasible time frame and these parameters will be discussed in the next chapter along with the relevant NMR physics.

The carotid arteries have an average diameter of 6.5 mm and 6.1 mm for males and females respectively [62]. This brings the in-plane region of interest section to a square with a size of about 7mm. Then the grid option over that area would determine the resolution which should be adequate for the accurate imaging of the vessel lumen and the reliable assessment of atherosclerotic plaque. Clinically approved MRI scanners, i.e. up to 3 Tesla, typically provide of the order of 1mm in-plane resolution, thus being inadequate for a detailed imaging. Currently, diagnosis is based on the experience of the radiologist, mostly via a database of images that have been correlated with the excised specimens under investigation. Figure 2.14 is a representation of how a resolution grid would appear over an average sized carotid artery. For instance, for a reliable assessment of the thickness of a thin fibrous cap (given earlier at $65\mu m$) a resolution of the order of $100\mu m$ would be valuable, although a higher resolution would be preferred. How this could be achieved will be discussed in the next chapter.

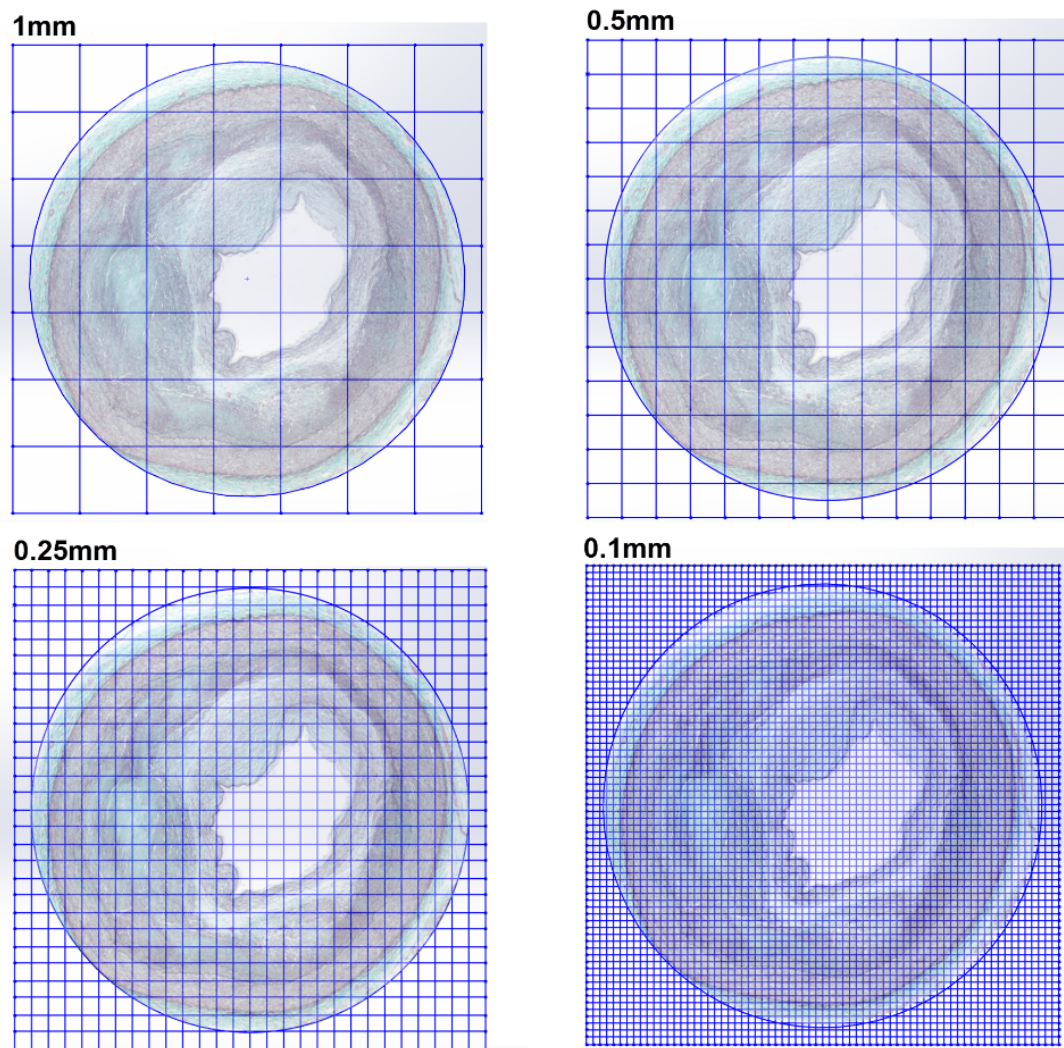


Figure 2.14: Resolution grid over an average atherosclerotic carotid artery. The 1mm in-plane is the real resolution achieved with high field scanners, 3 Tesla. Ultra-high field scanners, above 3 Tesla can improve the signal strength and images of 0.1 mm or $100\mu m$ have been acquired.

2.6 Summary

Atherosclerosis is an arterial disease mainly involving the large and medium sized systemic arteries, most commonly the carotid, coronary and peripheral arteries. The solidity of the fibrous cap has a key role in determining the rupture risk. Soft plaques seem to be more unstable since they have a large lipid core that is only separated from the lumen by a thin fibrous layer. These plaques are often considered to be vulnerable, posing a high rupture risk leading to severe thrombotic events. The bifurcations of carotid arteries are locations of high rupture risk due to the shear stress from the flowing blood. Furthermore, a rupture and clot separation can have critical consequences to the patient such as an ischaemic stroke.

Techniques such as magnetic resonance imaging (MRI) are important for the development of novel therapeutics as they allow for precise monitoring of the treatment effects over time. Identification of atherosclerotic plaque and early initiation of the treatment could be performed more reliably than the currently used invasive imaging modalities and have a positive impact on morbidity and mortality from atherothrombotic episodes.

MRI is currently the most promising tool for non-invasive imaging of carotid atherosclerotic plaque since it can image both the vessel lumen and the arterial wall. MRI can in principle provide contrast between a lipid-rich core and a fibrous cap and thus provide valuable information for the cap vulnerability. Perfusion imaging can also contribute to the assessment of the blood restriction in the brain due to stenotic arteries. However, the images presented earlier might appear difficult to interpret for a non-expert but improvement in the technology and the introduction of Ultra High Field scanners may be the key development to comprehend the mechanisms leading to atherosclerotic progression.

In the next chapter, nuclear magnetic resonance physics will be discussed and a detailed account of the technical and historical background of magnetic resonance imaging will be presented.

Chapter 3

Principles of Magnetic Resonance Imaging

3.1 Introduction

Nuclear Magnetic Resonance Imaging is a non-invasive modality which is based on the Nuclear Magnetic Resonance (NMR) property of matter. For instance, water molecules comprise two hydrogen atoms whose nuclei consist of one spinning proton, hence have an intrinsic magnetic moment which can be used for assessing water density distribution in biological tissues. The NMR property was known from the early years of the 20th century and the first technique utilising it was Magnetic Resonance Spectroscopy (MRS). It was developed in the late 1940s [63] and the composition of the sample could be assessed but not its density distribution.

Further characterisation from Bloch in 1946 [64] set the ground for all NMR applications. Imaging of biological tissues was an initial target application but it proved difficult to achieve due to the low performance of equipment at that time. Later, in 1960, the development of high static field magnets made spectroscopy possible in living systems. The first attempt to interact with biological tissues was the use of MRS in ^{31}P to measure pH of intact blood [65], and hence biochemical

MRS was introduced. Soon after, MRS in rat leg was performed in-vitro [66], and it became a clinical research tool in 1980 with Ackerman's first description of the surface coil [67].

The initial imaging experiments date back to the early 1970s, where a back projection reconstruction was performed on water-filled capillaries. The result was a 2D image formed from one dimensional density profiles acquired with the use of gradient coils [68]. Human imaging was first done on a finger [69], a hand [70] and a wrist [71], eventually leading to what MRI is today. In this chapter, a scientific background from underlying NMR property to 2D and 3D image acquisition will be outlined.

3.2 Nuclear Magnetic Resonance

The Nuclear Magnetic Resonance effect arises from the interaction between external magnetic fields and the magnetic moments of atomic nuclei in matter. Every nucleus component, proton or neutron, of an atom has an intrinsic angular momentum, commonly known as spin in the same way that the earth spins around its axis. Revolving electric charges are known, from Maxwell's equations, to create a magnetic vector. The spin of each particle is a quantified value and is expressed by integer and half integer numbers [72]. Depending on the spinning direction it takes a positive or a negative sign. The magnitude of spin angular momentum is given by equation 3.1.

$$|p| = \hbar \sqrt{I(I+1)} \quad (3.1)$$

where $\hbar$ is Planck's constant divided by 2π and I is the spin quantum number which takes integer or half integer values [72].

In nuclei with an even number of protons and an even number of neutrons there is a negative spin for every positive spin and hence no net magnetic moment. However,

in nuclei with odd numbers of nucleons there is an excessive 'unmatched' nucleon that creates a net magnetic moment. An example of such a nucleus is the hydrogen atom which consists only of one proton that can have $\pm\frac{1}{2}$ spin. Since hydrogen can bind with oxygen to form water, it is very abundant in biological tissues and for that reason it has become the reference nucleus in MRI.

When an external static magnetic field $\mathbf{B}_0$ is applied, these nuclear magnetic vectors tend to align either in the same or in the opposite direction. When the alignment is with the external field the nuclei are in a low energy state whilst if it is against, the nuclei are in a high energy state. The equation that describes the number of protons aligned with the field versus those ones against is:

$$\frac{N_{parallel}}{N_{anti-parallel}} = \exp\left(\frac{\gamma\hbar B_0}{kT}\right) \quad (3.2)$$

where k is the Boltzmann constant, T is the temperature in Kelvin, B_0 is the external field in Tesla. $N_{parallel}$ is the number of nuclei aligned with the field and $N_{anti-parallel}$ is the number of nuclei that are non-aligned.

The difference between the portion of nuclei that align with and against the external field is extremely low, of the order of 0.0005%. For instance, a magnetic field of 3T will force only about 10 more protons per million parallel to the field than anti-parallel. However, because tissues are comprised mainly of water there is a very large number of proton nuclei which overall give a detectable signal. Thus, the smaller the field, the less the net magnetization in the sample and the poorer the SNR. This is the main reason why stronger B_0 fields are desirable.

3.2.1 Larmor Precession

When the nuclei are aligned with an external static magnetic field, apart from their spin, they also maintain a constant motion similar to a gyroscopic spin under the gravitational force. In simple terms, the trajectory of the magnetic moment

Table 3.1: Gyromagnetic ratios of active NMR nuclei [73].

Nucleus	1H	13C	19F	31P	21Na
Gyromagnetic Ratio (MHz/T)	42.58	10.71	40.05	17.25	11.27

vector is a circle or in most cases an ellipsoid, see figure 3.1. This motion is called Larmor Precession and its frequency depends linearly on the B_0 strength, the Larmor frequency being given by:

$$\omega_0 = \gamma B_0 \quad (3.3)$$

where ω_0 is the angular frequency in rad/s, and γ is the gyromagnetic ratio measured in rad/T. For every nucleus there is a value of γ that describes its behaviour under static magnetic fields, see table 3.1.

In the absence of an RF excitation field there is no coherence, i.e. the spins are out of phase, to this precession and only the z component exists in a sample of fully relaxed spins.

3.2.2 Free Induction Decay

The principal idea in NMR is based on the fact that when nuclei are excited by electromagnetic waves oscillating at the Larmor frequency there is significant energy exchange with the spins and a portion of them changes energy state. In particular, some nuclei go from the low energy to the high energy state or vice versa. After a certain time the radio frequency excitation ceases and the previous equilibrium is restored under the effect of the strong static magnetic field B_0 .

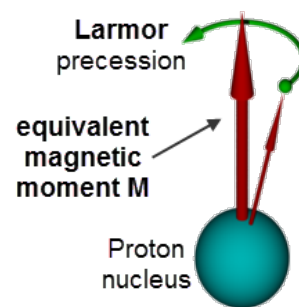


Figure 3.1: Larmor precession, apart from the magnetisation there is a precession as depicted by the green arrow.

During the 'restoration' period the magnetic moments change direction to return to

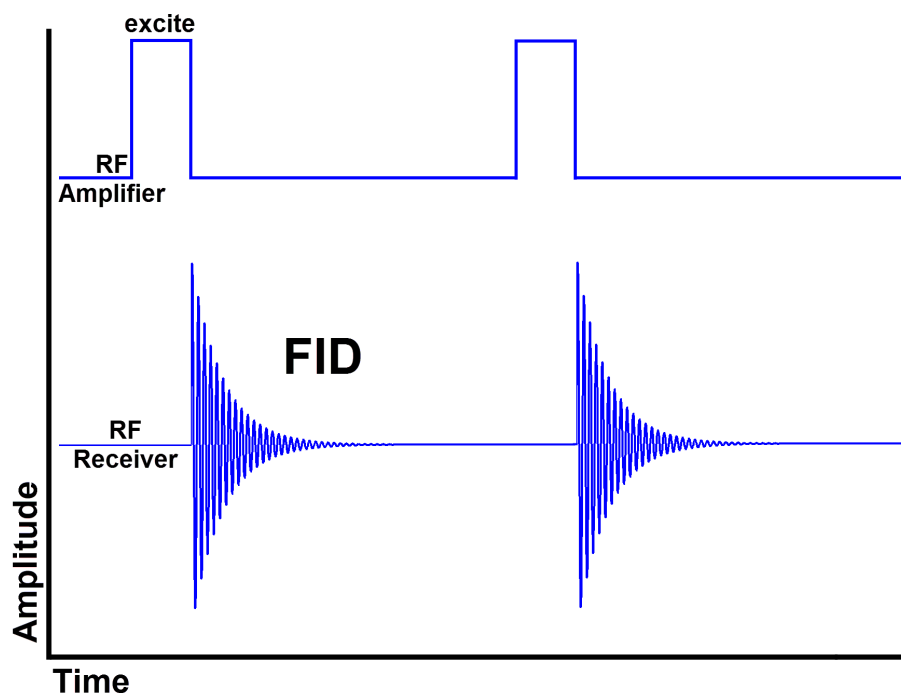


Figure 3.2: Free Induction Decay under the influence of spin-spin relaxation (T_2) constant. On the left is depicted the FID signal and on the right the corresponding useful demodulated 'envelope'.

their equilibrium condition. Changing magnetic vectors cause a change in magnetic flux which in turn can induce currents according to Faraday's law. However, the useful signal in NMR comes from the transverse rotation only; since the Larmor frequency lies in the RF bands, the MR signal can be obtained and demodulated by radio frequency instrumentation. The signal arising from a spin relaxation experiment is called Free Induction Decay (FID) and is shown in figure 3.2.

A very useful and commonly used modelling of spin excitation is the use of tipping magnetisation vectors. Each time an RF pulse is applied a magnetic spin that initially is aligned with $\mathbf{B}_0$ and coincides with the z -axis of a Cartesian system is tipped on the xy plane, figure 3.3, and additionally acquires the same phase as the external field. Right after the RF pulse all the excited spins are tipped on the xy plane and, having the same phase, a signal is generated rotating on the xy plane and around the z -axis.

The intrinsic signal is rotating as previously explained and is detected on the xy

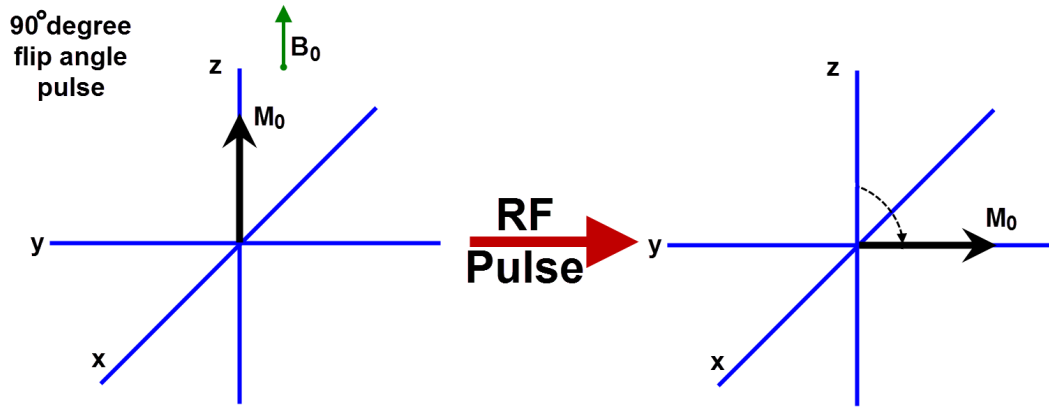


Figure 3.3: M_0 vector representing the spin magnetic moment, initially aligned along B_0 on z -axis and tipped on the x - y plane after a 90° excitation.

plane only. In order to excite the spins the external RF field must also be rotating on the xy plane with the Larmor angular frequency ω_0 . This rotational field is called B_1 and is the useful excitation field generated or received by the RF coils. The B_1 field can be broken down to the B_1^+ and the B_1^- components where the first represents the field that rotates in the same angular frequency direction as the precessing spins and the second represents the field in the opposite angular direction. The B_1^- is in essence the receive sensitivity field of the RF coil since it follows the ‘unwinding’ of the spins after the excitation pulse. Both fields will be a matter of detailed discussion later in the study.

3.2.3 Magnetisation Relaxation

The restoration to the equilibrium state, or relaxation, is described by two time constants, T_1 and T_2 . The Bloch equations [64] describe the way the system gradually returns to the pre-excitation state. T_1 is called the spin-lattice relaxation time and it corresponds to the time constant needed by the net magnetization vector to revert to an alignment with the static magnetic field by a $\frac{1}{e}$ factor. The rate of recovery of longitudinal magnetization, M_z , is described by the differential equation:

$$\frac{dM_z}{dt} = -\frac{M_z - M_0}{T_1} \quad (3.4)$$

where M_0 is the equilibrium magnetization along the $\mathbf{B}_0$ direction.

The second time constant, T_2 , is called the spin-spin relaxation time and it reflects the exponential decay of the transverse magnetization after excitation. Once excited, nuclei have the same phase but the exchange of energy between spins causes gradual de-phasing. The decay of transverse magnetization is given by:

$$\frac{dM_{xy}}{dt} = -\frac{M_{xy}}{T_2} \quad (3.5)$$

Typical biological values of T_2 range from 40 to 200ms and T_1 is roughly 10 times bigger [73]. Another constant used often is T_2^* . Spin-spin relaxation does not take in to account the inhomogeneities in the $\mathbf{B}_0$ field that stem from the insertion of subjects with variable magnetic susceptibilities in the static field. The latter effect leads to slightly different Larmor precession between spins that are positioned in different magnetic field environment and thus dephasing is faster than what it would be if only T_2 dictated the decay. For this reason the relaxation is faster and T_2^* is always smaller than T_2 and in fact dominates the NMR signal loss in most cases.

3.2.4 Flip Angle

An important concept in NMR parametrisation is the flip angle α . In simple terms, the longer the RF field is applied, the more the nuclei will change their energy state. The flip angle expresses the relation of the up/down spin ratio before and after the excitation. It is a way to measure the strength of the excitation, and it is estimated by:

$$\alpha = \gamma B_1 \tau \quad (3.6)$$

where τ is the pulse length. Depending on the units of γ the flip angle can take units of cycles, rads and, most often, degrees.

The flip angle concept applies very well to the spin vector model as explained before and in fact the angle between the magnetisation vector before, ie. in the equilibrium state, and after the RF pulse represents the flip angle. Thus, if the M_0 is tipped to the xy plane the flip angle is 90° . In most NMR experiments the excitation varies from as low as 10° to high power excitation schemes of 90° and multiple 180° pulses. More details will be given in section 3.4.

So far, the principles of NMR have been described. By setting the flip angle, α , and acquiring signal under T_1 , T_2 relaxation using spectroscopy alone, a particular substance in a sample can be detected. However, it is impossible to detect where this substance is within the sample.

3.3 Magnetic Resonance Imaging

For MRI, spatial encoding of the NMR signal is introduced in order to extract images. In 1973, Lauterbur proposed the use of a field gradient to localise the NMR signal [68]. A field gradient was applied in a series of different directions, each creating a projection through the sample. The projections were then combined to form an image using back-projection, a method used in CT scans [74]. In the same year, selective excitation was proposed by Mansfield et al. [75] with the help of a gradient parallel to the main static field; the same method is still used today.

The first Fourier imaging method was introduced in 1975 by the Ernst group [76]. A 2D phase and frequency encoding readout was used but the equipment of the time did not allow for processing full volume datasets and hence 3D imaging. In 1977 Mansfield [77] introduced a technique for fast imaging, the so-called Echo Planar Imaging (EPI) that opened new horizons to the capabilities of MRI, most notably the advent of fMRI scans. Throughout the years, computational power has

increased making it possible to perform complex image processing and analysis as well as handling large data sets. Along with this development, MR equipment has evolved rapidly rising from 0.1 Tesla to ultra high field scanners of 7 Tesla and beyond.

In short, the main differentiation of MRI compared to NMR is the ability of the former to extract cross-sectional images of a sample. This is done in two generic stages; first slice selection and excitation and the second stage comprises the spatial encoding of the one dimensional NMR signal. The first stage is described next whereas the second stage, being far more complex, is outlined in section 3.4.

3.3.1 Slice Selection and Excitation

From the Larmor equation (3.3) it is clear that hydrogen nuclei precess at different frequencies for different external fields. If the static field can be modified, the Larmor frequencies can also be altered. The idea of Lauterbur and Mansfield [68, 69, 78] was to apply a gradient magnetic field which would make the Larmor frequency dependent on its position, see figure 3.4. Thus, an RF pulse set to the appropriate frequency would excite only the area in the sample with the corresponding static field. Then a thin slice can be deliberately excited or ‘selected’ and the subsequent FID will only arise from that area. By obtaining several slices sequentially, the whole Region of Interest (ROI), such as a full brain, can be covered and an assessment per slice or a 3D reconstruction of the whole volume can be made.

From the Larmor frequency equation:

$$\omega_{grad} = \omega_0 + \delta\omega(z) = \gamma(B_0 + z \cdot G_z) \quad (3.7)$$

where z represents distance from the isocentre along the Z axis in figure 3.4. The G_z is the gradient slope, usually up to about 50 mT/m; it is also called the Slice Selection gradient and usually denoted as G_{SS} in pulse sequence diagrams.

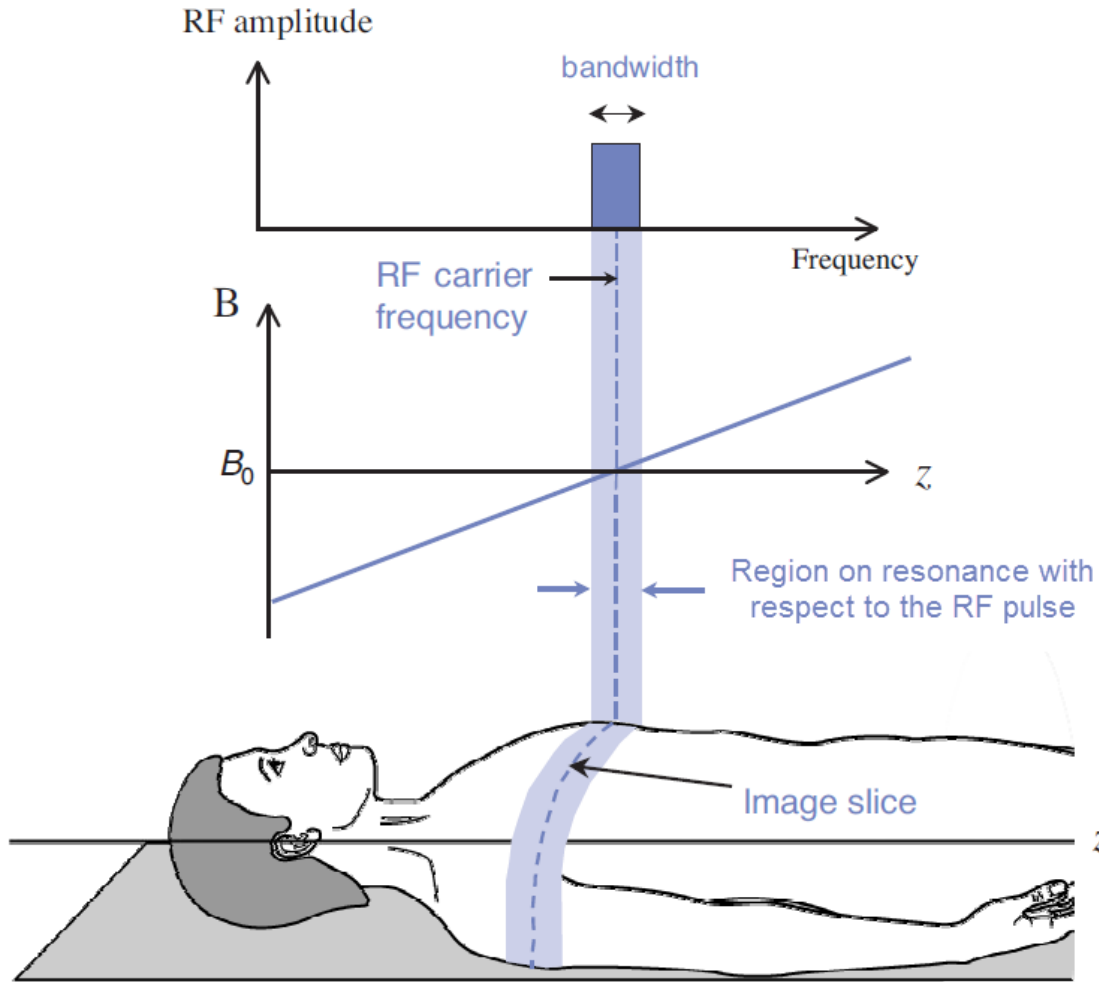


Figure 3.4: Slice excitation with Z gradient field. On top the RF sinc pulses generate a square frequency profile, ideal for exciting only a certain thickness around the centre frequency (ω_0) as shown on the bottom. Reproduced with permission from [79]

Slice excitation is often done by RF pulses of sinc shape (in time) due to the fact that they have a square profile in the frequency domain. The latter enables the selection of a thin slice (narrowband pulse) as shown in figure 3.4. There are two main excitation methods, spin echo (SE) and gradient echo (GE) figure 3.5. In spin echo, spins are excited by a 90° pulse and after a short time an 180° pulse is applied to flip the phase of the spins leading to a subsequent signal peak (or echo). A Fast Spin Echo (FSE) or Turbo Spin Echo (TSE) sequence is a spin echo followed by successive 180° pulses in order to refocus the nuclei successively until the intrinsic

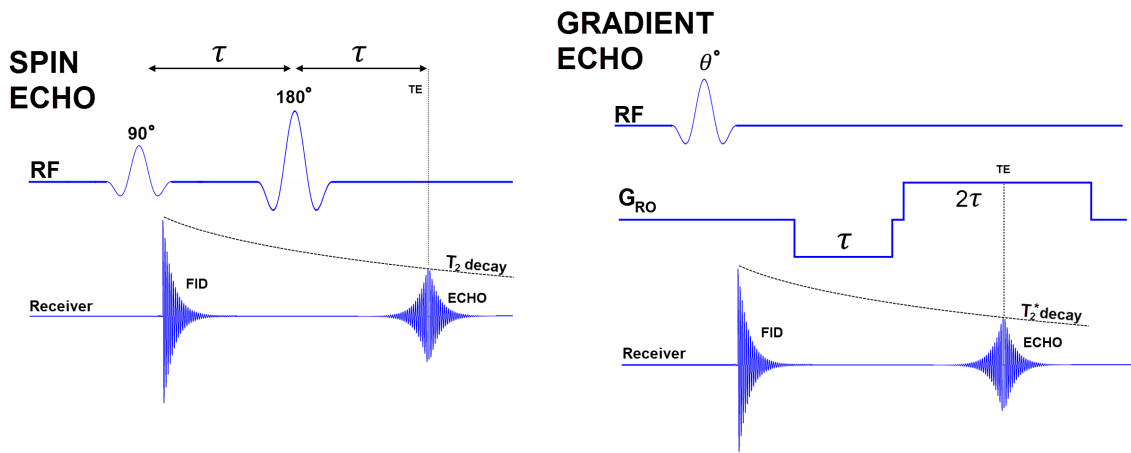


Figure 3.5: Spin Echo (left) and Gradient Echo (right) excitation sequence. During the RF pulses a SS gradient is applied (not shown here).

T_2 signal weakens to the point where it is no longer retrievable, see figure 3.6.

In gradient echo sequences the RF pulse is usually weaker than 90° and the mechanism of ‘echo’ generation is via the application of an equal and opposite (or negative) gradient along the readout direction (following a preceding readout dephaser gradient). The spins under the influence of the negative gradient start to precess counter circularly to the positive gradient and will eventually refocus producing an echo, figure 3.5(right).

After slice excitation with the methods described, the NMR signal (FID in figure 3.2) is then acquired under the presence of gradient fields that are aligned with the transverse plane, xy . The FID then deviates from what it would be without any gradients applied and, its envelope includes encoded information of the proton distribution in the slice. In particular, the additional signal decay derives from the spin de-phasing caused by the slightly different precession due to the gradient presence, see figure 3.7.

Before describing the spatial encoding of the NMR signal, it is useful to introduce the various MRI contrasts and how they are achieved.

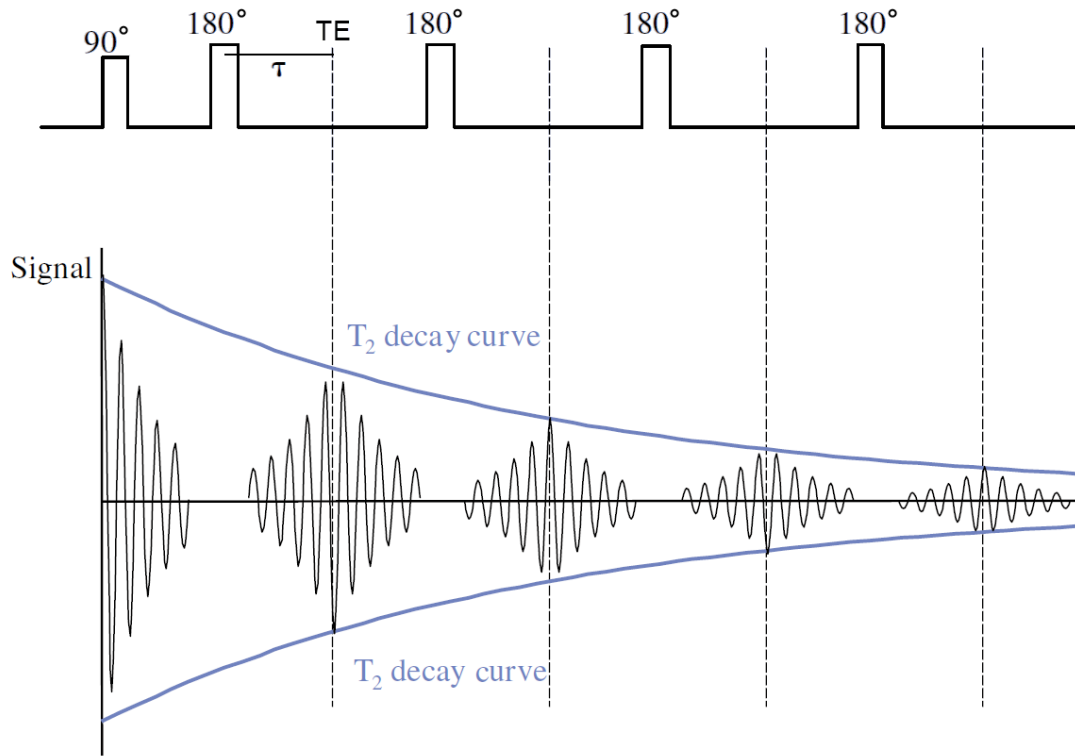


Figure 3.6: T_2 decay curve for fast spin echo imaging, a measurement with successive flipping of the spins until the T_2^* relaxation (exchange of energy between spins and inhomogeneities of the B_0 field) completely erase the signal. Reproduced with permission from [79].

3.3.2 Image Contrast

The advantage of MR imaging is that the image contrast is not standardised but it can be manipulated through different imaging techniques. By adjusting the timing and the intensity of the sequence elements it is possible to alter the contrast and make, for instance, body fluids look darker or brighter relative to surrounding tissue. CT scans, on the contrary, depend only on the X-ray absorption during propagation in the tissues and hence in the absence of injected agents the contrast cannot be manipulated. This property of MRI is extremely useful in facilitating the diagnosis of certain conditions such as oedema etc.

In order to create a contrast three main mechanisms are utilised. One is the longitudinal relaxation dictated by the T_1 time constant, equation 3.4, and the

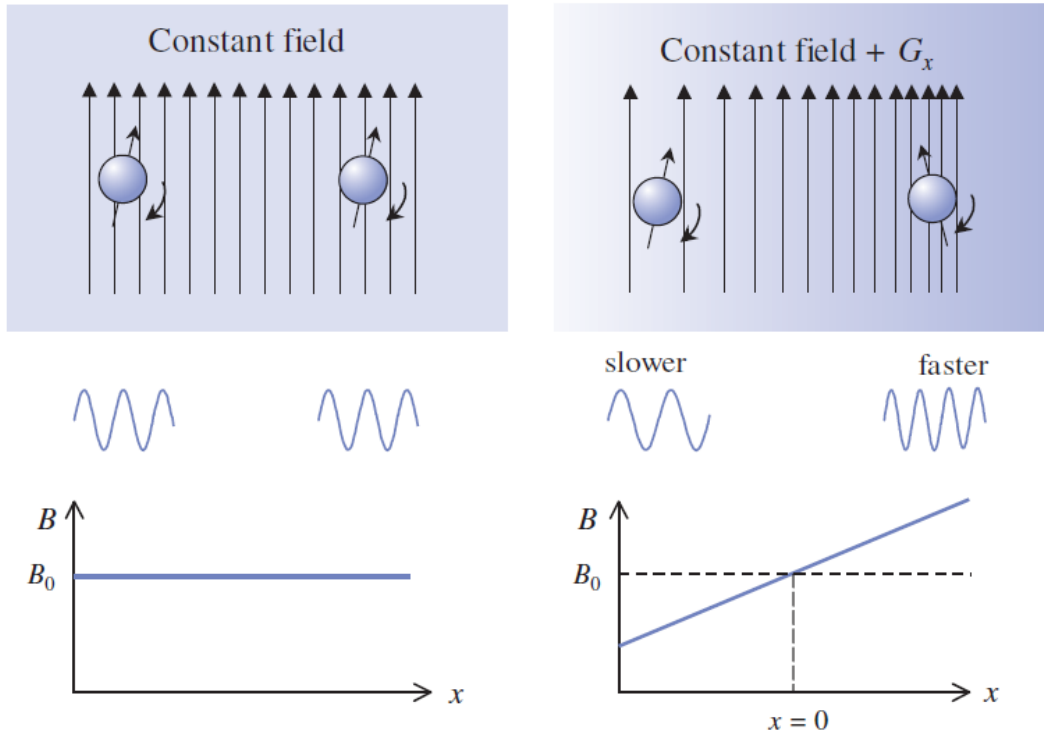


Figure 3.7: On the left no gradient is applied and the B_0 is the same along the two spins which precess at the Larmor frequency, ω_0 . On the right a gradient is applied, usually with a slope of the order of 10 to 50 mT/m forcing spins that lie in different places to precess with a slightly different Larmor frequency $\omega_{grad} = \omega_0 + \gamma \cdot G_{grad} \Delta(x)$. Reproduced with permission from [79].

other is the transverse one described by the T_2 time constant, equation 3.5. The third contrast mechanism based purely on the proton density distribution and is abbreviated PD.

T_1 Weighting

T_1 weighted imaging can be generated since different tissues have different T_1 time constants. Also, the values of T_1 are always longer than the values of T_2 thus making it possible to acquire FIDs before spins re-align with the $\mathbf{B}_0$ field. The fact that there is remnant longitudinal magnetisation makes the FIDs have different intensities based on the rate of T_1 relaxation for each tissue and thus the contrast is created.

For spin echo excitation the TR is made short in order to bring the T_1 relaxation to a steady state and as a result to highlight the T_1 differences that stem from each

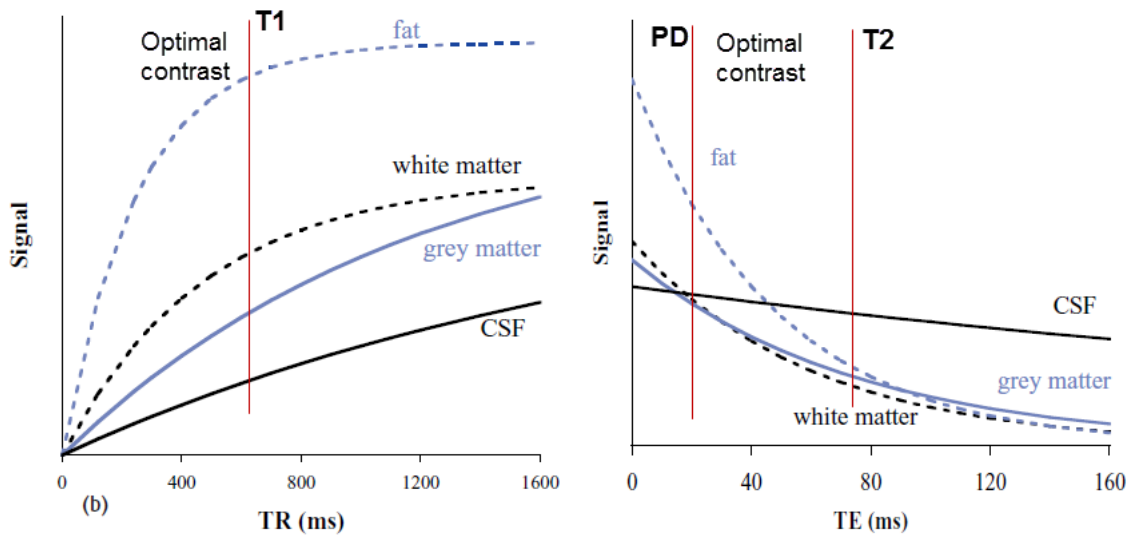


Figure 3.8: On the left, the longitudinal (or T_1) relaxation is sketched for each tissue, i.e. fat; white matter; grey matter and cerebrospinal fluid (CSF). For T_1 weighted imaging the choice of optimal contrast is marked with a red line, that is a TR of about 500ms. On the right diagram, the transverse (or T_2) decay curve is drawn for the same tissues and the first red line marks the TE selection for the optimum contrast for proton density imaging; the second line marks the optimum TE for T_2 weighted imaging. It can be observed that for T_2 weighted scans the contrast between solid tissues is lost and fluids like CSF give a higher signal. Adapted from [79].

tissue type providing the contrast, see figure 3.8. The TE is selected depending on the sequence but it is usually set to a short value to limit any T_2 effect in the contrast.

The T_1 weighted scans depict fluids as dark, tissues with mid grey and adipose tissue as bright. T_1 weighted images are often called anatomical scans as they show very clearly the boundaries between tissues. T_1 weighted images can also be generated with GE sequences when again the T_1 base signal level is brought to a steady state with a relatively large flip angle, a short TR and time echo due to the GE nature. Table 3.2 provides a thorough outline of how TR, TE and flip angle affect the contrast.

T_2 and T_2^* weighting

T_2 weighted contrast is generated by using long repetition times (TR) so that the longitudinal magnetisation is completely recovered and an appropriate, relatively long, echo time (TE) is used for capturing the maximum possible T_2 contrast among tissues, see figure 3.8 (right). The fluids in T_2 weighted images have a bright appearance with tissues and fat showing in mid-grey. These images are also called ‘pathology’ scans due to their sensitivity to fluid accumulations that usually mean injury or oedema.

As explained in section 3.2.3, the actual transverse signal decay of the tissues is faster than the theoretical value due to susceptibility effects and $\mathbf{B}_0$ inhomogeneities, figure 3.9. However, based on the type of excitation the effect of $\mathbf{B}_0$ inhomogeneity can be less apparent producing ‘pure’ T_2 images, or intentionally included to generate T_2^* contrast. In particular, when the excitation scheme is a gradient echo the contrast is T_2^* but for spin echo excitation the contrast is plain T_2 -weighted. This happens because GE relies on the $\mathbf{B}_0$ distribution via a strong stable gradient to re-focus the spins amplifying thus the effect from the imperfect field. In SE the spins are refocussed by both a gradient and by a strong 180° degree RF pulse that has the effect of refocussing any imperfections in the $\mathbf{B}_0$. The contrast and greyscale appearance of tissue types is broadly the same for T_2 and T_2^* scans.

Proton Density weighting

The intrinsic contrast arising from the water density variation among the tissues is not significant and hence not very useful clinically apart from a few occasions (nerve roots in cervical spine, oedema). In general, the shorter the TE, the more PD contrast is produced and a long TR is always required in order to eliminate any T_1 relaxation effects.

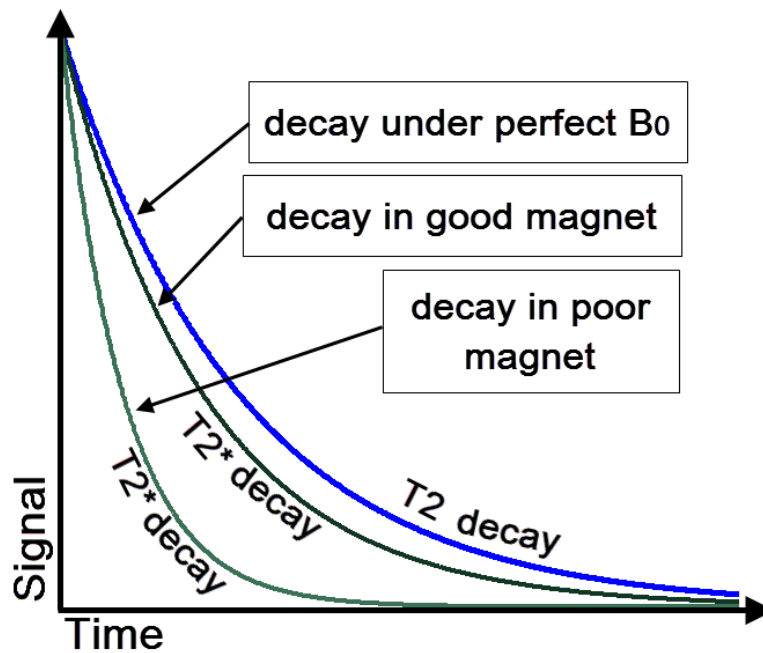


Figure 3.9: The level of T_2^* deviation from the true T_2 depends on the quality of the B_0 distribution. To some extent the static profile is corrected by the shim coils adding or subtracting magnetic field and compensating for the inhomogeneities.

Spin Echo sequences		
TR	TE	
	Short, less than 40ms	Long, more than 75ms
Short, less than 40ms Long	T_1 weighted	Not useful
Long, more than 1.5 sec	PD weighted	T_2 weighted
Gradient Echo sequences		
Flip angle , α	TE	
	Short, less than 15ms	Long, more than 30ms
Small, less than 40°	PD weighted	T_2 weighted
Large, more than 50°	T_1 weighted	Not useful or fMRI

Table 3.2: The typical parameters used for different contrast are outlined for a spin echo excitation sequence (top) and for the gradient echo ones on the bottom table.

3.4 Pulse Sequences and Image Formation

So far an explanation has been given on how a slice is selected and a description has been given of the contrast manipulation that MRI can offer to the radiologist. In this section a description is provided of how the 1 dimensional NMR signal is spatially encoded in such a way to construct 2D or 3D images. Also, the methods of spatial encoding using the k-space matrix are described.

3.4.1 *K*-space

In MRI, pixel intensities are not directly obtained as is done in X-ray or in CT scans, where multiple X-rays are combined for a back-projection reconstruction. In MRI the data are gathered in the spatial frequency domain meaning that the intensities captured depict how much the image fluctuates in space, measured in cycles per metre. The equivalent unit in electromagnetic theory is the wave number, or k , calculated as $2\pi/\lambda$ and its units are rad/m . This k number is what is obtained in MRI, but the difference is that it is obtained in two dimensions and for a big enough dataset to include the spatial variation of human anatomy, i.e. about 1000 cycles/m. The platform on which these numbers are stored is a matrix called k -space because it holds all the wavenumbers, k , of the spatial frequencies of our image. Once enough data is gathered the image can be re-constructed using a 2D Fourier transformation, see figure 3.10, [75, 80].

K -space, as shown on the left in figure 3.10, has two Cartesian axes, k_x and k_y , that intersect in the middle of the matrix. The intensity at any coordinate combination, signal acquired by a scan, quantifies how much of this particular spatial frequency is included in the image. Thus, coordinates close to the zero crossing include low spatial frequencies, i.e. rough details of an image, and large coordinates far from the centre include high spatial frequencies that represent the fine details, figure 3.11. The image then can be re-constructed via a 2D Fourier transform.

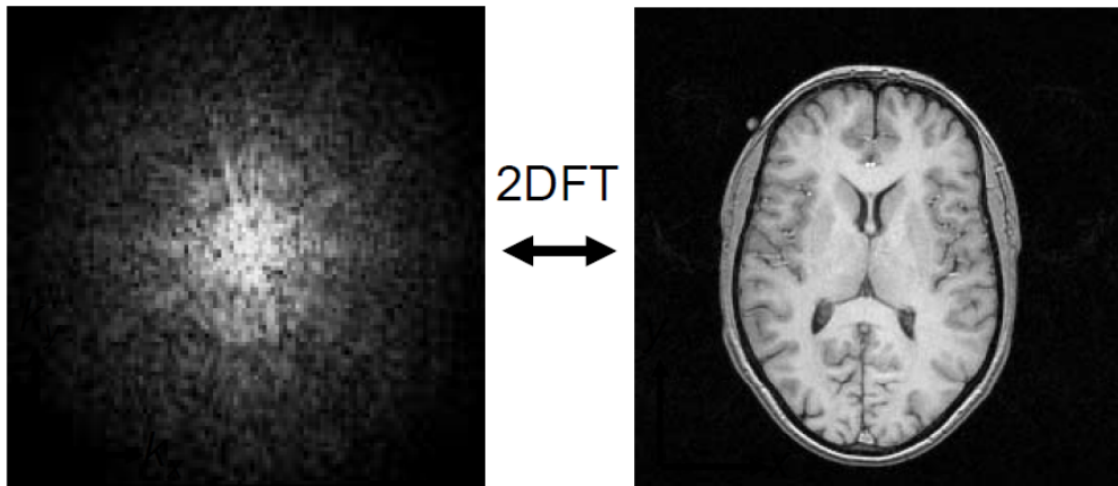


Figure 3.10: Measured Signal arranged in k -space matrix (left) and the reconstructed image after 2D Fourier transformation (right).

The k -space is a hermitian conjugate symmetrical matrix meaning that what appears in one quadrant will appear hermitian conjugated in the diametric one. If k -space is considered a complex number system, then the conjugate of any number on k -space is present too. Due to this it could be assumed that there is no need to acquire the whole k -space but just half of it and fill the rest. Such technique is known as ‘half Fourier’ allowing for smaller scan duration. A similar technique whereby half of the echo (‘half echo’) is acquired is also a partial k -space method. However, even if they reduce the scan time these are lossy techniques due to phase errors in the MRI data due to the $\mathbf{B}_0$ imperfections as discussed earlier. Other factors that can induce asymmetry in k -space are motion, physiological activity, and coil properties.

There is theoretically a perfect k -space for every image and for biological imaging that can go down to roughly $10\mu m$ voxel dimension, but in practice there are many limiting factors and the most prominent is the strength of the intrinsic MR signal versus the white noise found in electronic components (signal to noise ratio, SNR). There are methods to improve the SNR of a scan by repeating the acquisition and working on sequence methods but the most important factor in the SNR equation is the acquisition device or the RF coil which will be discussed extensively later.

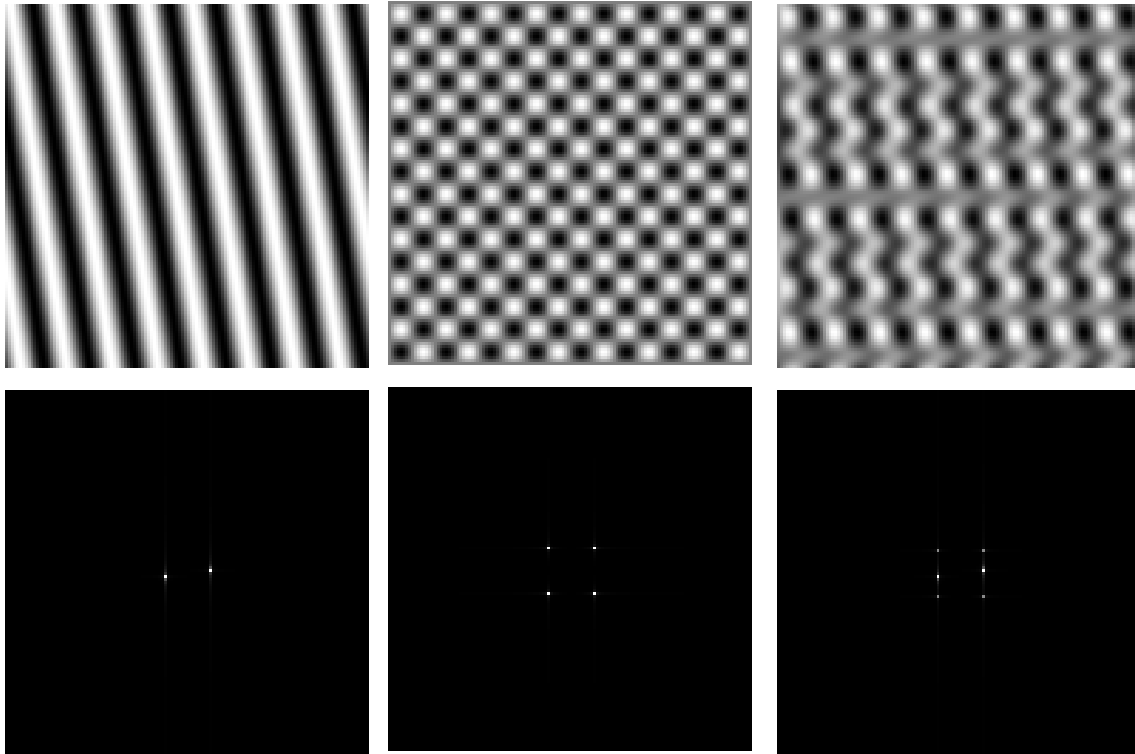


Figure 3.11: On top is the image and on the bottom the corresponding k-space. In the first example there is only one oblique spatial frequency, close to the axes centre, giving two dots in k-space due to the symmetry. In the second image two spatial frequencies with the same intensity create a perfect chessboard. As the k-space becomes more complicated, in the third example, intuitive understanding of the generated image is more challenging.

The question now is how the scanner determines where the signal comes from in space since what is received is just a 1D wave originating from the whole sample or subject. The answer is Spatial Encoding and this will be discussed in the next section.

3.4.2 Spatial Encoding

Frequency Encoding

During acquisition the signal is spatially encoded with magnetic field gradients which relate any point on the imaging plane to the local Larmor frequency. The signal comes from the whole excited slice (or volume) and its constituent frequencies can be extracted by a Fourier transform. As discussed before, in the absence of gradients

during acquisition the signal would be an FID and the frequencies extracted would be the characteristic peaks of the sample components (spectroscopy). By the gradient application, the intensity of any component, seen as one peak in spectroscopy, would be spread out providing information of its density along the gradient direction. This spatial info is extracted by the FT of the MR signal showing that the latter includes the spatial frequencies of the image.

The method described in the previous paragraph is called frequency encoding (FE) and is by convention employed along the x -axis of the image plane. In FE a gradient is applied during data acquisition and forces nuclei to precess at different frequencies in order to utilise the localisation mechanism already described. The FE gradient, also referred as read-out (ro), is similar to the slice selection one but in an orthogonal direction. Figure 3.12, on the left, illustrates the concept behind FE.

Phase Encoding

For the other dimension it is not possible to use the same mechanism due to interference with the FE gradient causing different locations to be represented by the same frequency. Thus, for the second dimension, by convention taken to be the y -axis of the image, Phase Encoding (PE) is used. In PE, constant duration gradients with varying amplitude are applied just before the FE gradient and the Analog to Digital Converter (ADC) window. The idea behind PE is to add phase to the precessing nuclei so that they accrue a phase offset just before the read-out gradient. The phase added can go from -180° to 180° for the whole k -space and is added incrementally for every PE line.

The phase difference between PE lines is in fact a phase differential over space $d\phi/dy$ which in effect represents frequency change since frequency is phase accumulation over time, $\phi = 2\pi ft$. The phase addition is also symmetrical ranging from -180° to 180° degrees and adjusted for the selected phase encoding steps, see figure 3.12. Phase encoding steps are generally powers of 2 to facilitate the computations

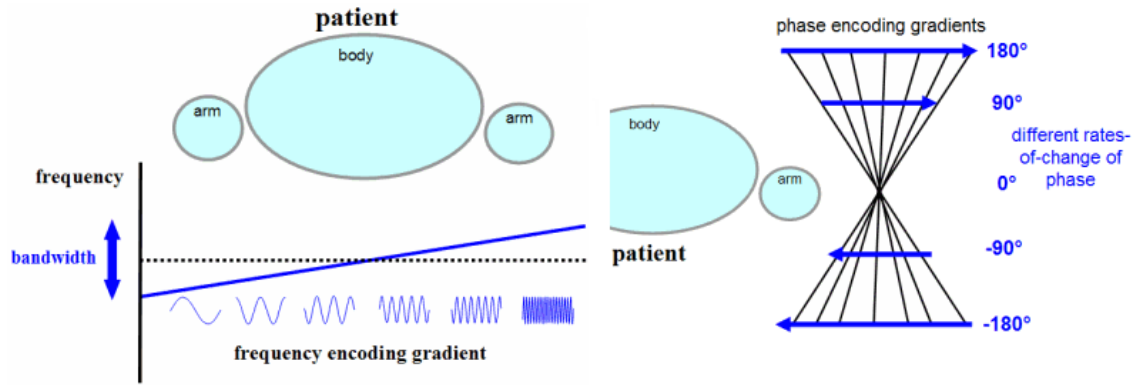


Figure 3.12: Frequency encoding method (left). The applied gradient makes each vertical ‘stripe’ along x -axis resonate at an increment frequency making it possible to differentiate the position based on the received spectrum. In phase encoding (right), before each acquisition the phase of the spins is altered by a short duration gradient for each line making it possible to differentiate the position along the second axis once all data is gathered. Reproduced with permission from [81].

and usually they take values such as 64, 128, 256. The next stage to be considered is the receiver and the sampling rate of the signal.

3.4.3 Receiver Bandwidth

The MR signal is amplitude modulated (AM) at the Larmor frequency so after going through the necessary amplification it is demodulated and sampled with an analog to digital converter (ADC). The ADC window is centred around the echo acquiring a number of samples that corresponds to the FE encoding steps, aka base resolution and is selected by the user. The echo formation rate depends on how quickly the phase accrues over the duration of a gradient; hence, the FE gradient strength is the determining factor of the echo duration, i.e. refocusing and decaying spins. For strong gradients the echo will occur fast whilst for lower values it will spread out in time. The sampling period, T_s , is adjusted accordingly to fit the echo duration and thus a high read out gradient requires a short T_s and wide receiver BandWidth (rBW) where, $rBW = 1/T_s$, and a smaller gradient a lower rBW.

According to Nyquist [82, 83], for the reliable digitisation of an analog signal, a sampling frequency, f_s , of at least double the highest frequency in the signal,

called the Nyquist frequency (f_N), is required. The ADC can operate at different sampling rates and whilst in theory a $2 \times f_N$ is required, in practice oversampling is utilised for better performance of low pass filters that are applied during the image reconstruction. The sampling rate is calculated by the scanner after the setup of the sequence parameters in order to include the required frequency content. The Field of View (FoV) is a parameter that can be modified and affects the read-out gradient strength too; the FoV combined with the FE steps and the rBW set the image resolution since pixel size = FoV/FE steps. For instance, 200mm FoV and 256 FE steps we get a 0.78mm pixel size. For the determination of the voxel size the PE steps are set relative to the FE steps in % and the slice thickness is set separately, see section 3.3.1.

There are SNR considerations when setting the BW; a wider BW leads to more noise and on the contrary, a smaller BW will exclude fine detail but will be less noise sensitive, see table 3.3 for a summary of how parameters change when handling the BW. Equation 3.9 shows how the SNR changes relative to rBW, FoV and a given matrix size with N_x FE steps and N_y PE steps.

	Lower Bandwidth	Higher Bandwidth
Pros	Higher SNR	Less Chemical shift
	Less noise	Shorter min. TE/TR
	Smaller minimum FOV	Shorter echo spacing
		Reduced susceptibility artefacts
Cons	More chemical shift	
	Longer min TE/TR	Lower SNR
	Longer echo spacing	More Noise
	Increased susceptibility artifacts	Larger minimal FOV
	More metal artifacts	

Table 3.3: Summary of the advantages and disadvantages over selecting lower or higher bandwidth. A low BW can exemplify susceptibility artefacts and requires longer TE and TR. At higher BW the most prominent adverse effect is the degradation of SNR.

The relation between the SNR and the selected BW, excluding the hardware

factor that is accounted for in the K constant, can be summarised as:

$$SNR = K(\text{voxel volume}) \sqrt{\frac{\text{number of measurements}}{\text{bandwidth}}} \quad (3.8)$$

$$SNR = K \cdot \left(\frac{FOV_x}{N_x} \cdot \frac{FOV_y}{N_y} \cdot \Delta z \right) \cdot \sqrt{\frac{N_x \cdot N_y \cdot NEX}{BW}} \quad (3.9)$$

where FOV_x and FOV_y are the selected field of view for the x and y axes, N_x , N_y are the number of FE and PE steps respectively, Δz is the slice thickness, NEX is the number of averages and BW is the receiver bandwidth. Scanners use this equation to give the operator a relative SNR improvement or reduction depending on the modifications performed in the sequence.

As far as the K factor is concerned, it includes the field strength dependent factors such as the pulse sequence parameters (TE, TR), sample properties (proton density, T_1 , T_2) and the hardware dependent factors such as coil sensitivity profile, matching and tuning to the load, pre-amp quality and performance (noise figure, decoupling) and noise levels. The last four will be extensively discussed and optimised in the next chapter.

3.4.4 Pulse Sequence Trajectory

The sequential events including RF excitation pulses, gradient field switching and ADC acquisition windows form the pulse sequence. The way k -space is filled with the raw data depends on the sequence played out each time. The sequence specifications can affect the quality of the image, the acquisition speed and manipulate the contrast as discussed earlier.

Image readout is determined by the gradient fields which define a trajectory of acquisition within Fourier space, being defined in [68] and in equation (3.11), for any combination of G_x , G_y , G_z as shown below.

$$G = \begin{pmatrix} G_{SS} \\ G_{FE} \\ G_{PE} \end{pmatrix} = \begin{pmatrix} G_z \\ G_x \\ G_y \end{pmatrix} = \begin{pmatrix} dB_z/dt \\ dB_x/dt \\ dB_y/dt \end{pmatrix} \quad (3.10)$$

The fact the first terminology is preferred is because in real subject scans, the gradient orientation rarely coincides with the ‘convenient’ Cartesian system. Oblique slices can be selected and FE and PE gradients are constituted by the three, X, Y and Z gradient coils.

$$k(t) = \gamma \int_0^t G(\tau) d\tau \quad (3.11)$$

where $G = [Gx, Gy, Gz]^T$ represents the three gradient vectors along each axis. Thus the trajectory defined by $k(t)$ shows the recording path from the one dimensional time course signal to two or three dimensions in k -space.

The acquisition trajectory can take various shapes and thus sequences have developed depending on the specifications of the MRI application. Examples include the standard Spin-warp sequence [84], Spiral imaging [85], burst trajectories [86] and others. A sequence that enabled MRI to go a step further in applications such as cardiac or functional imaging is Echo Planar Imaging (EPI), introduced by the Mansfield group [77]. In EPI, a full slice can be acquired during one excitation meaning that sub-second images can capture organs with significant temporal variation such as the heart.

In figures 3.13, 3.14 and 3.15, a time axis is drawn on the left. The RF axis represents the timing of sinc pulse transmission, G_{sl} , G_{fr} and G_{ph} represent the X, Y and Z gradient components in time. On the right a slice of the k -space sampling trajectory is presented. Since it is a two dimensional representation the trajectory is drawn by two of the gradients in each picture and the third one is used to produce the slice selection.

Cartesian Imaging

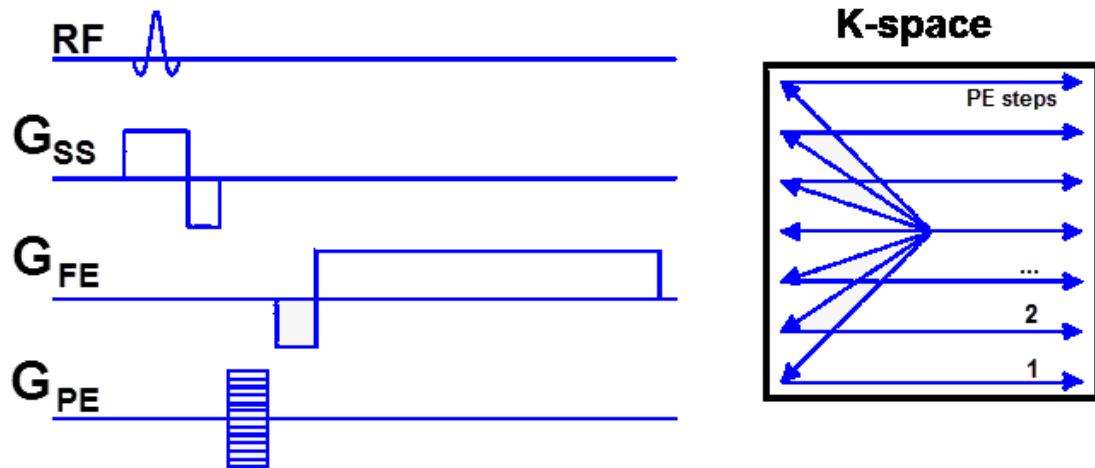


Figure 3.13: Standard or Cartesian imaging sequence gradient layout for one line on the left and the k -space filling trajectory on the right. Adapted from [73].

Spiral Imaging

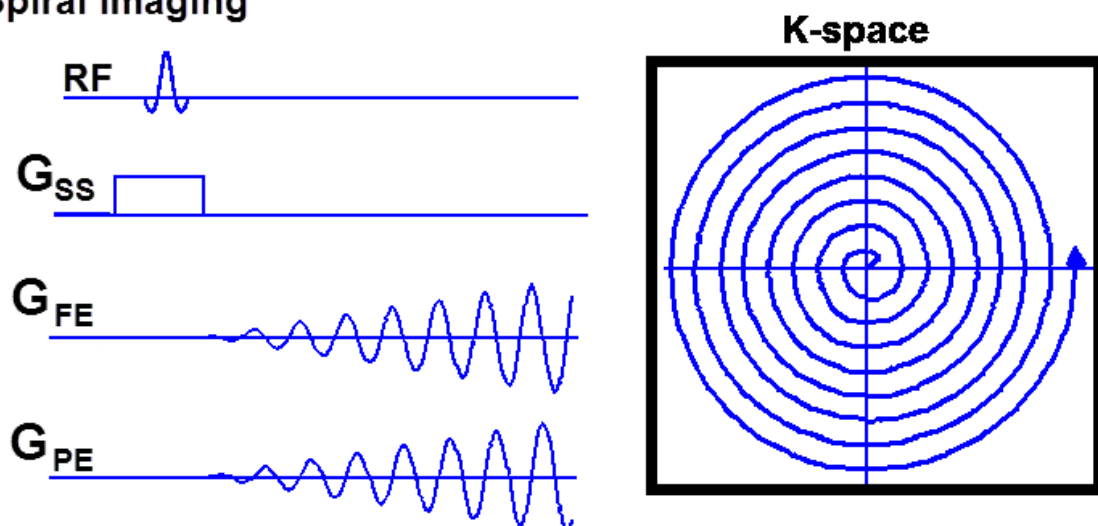


Figure 3.14: Spiral Imaging sequence and k -space trajectory. Adapted from [73].

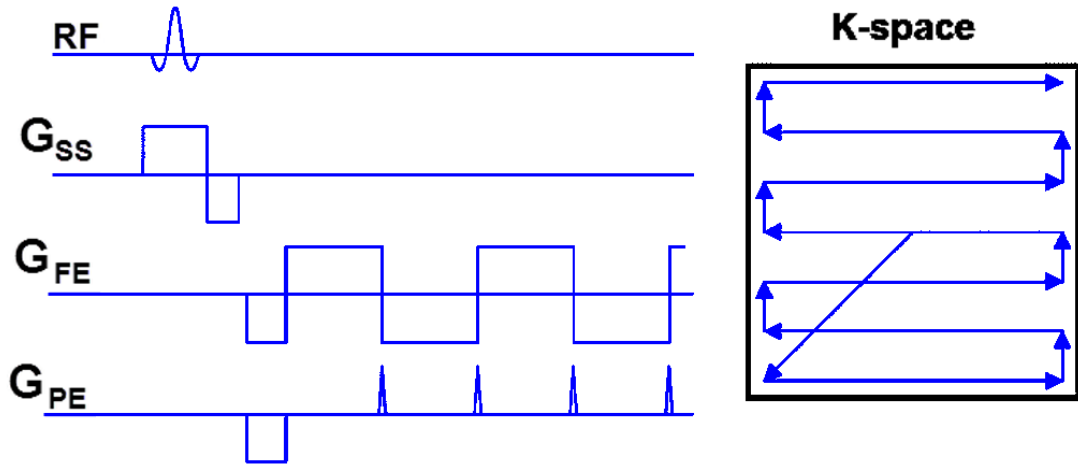
EPI Imaging

Figure 3.15: Echo Planar Imaging sequence (left). After slice selection via the RF pulse and the z gradient (G_{sl}), the k -space is filled by quick successive readouts (right) in effect producing a relatively still image of the specific slice. Adapted from [73].

3.4.5 Functional Imaging

Aside from the more ‘traditional’ T_1 , T_2 and T_2^* images, it is possible to obtain contrast based on functions of the body. For example, this can be the level of brain activity based on external stimuli, or quantification of directionality of the brain axons via diffusion imaging, i.e. water molecule movement in the tissues. Other techniques use a combination of tailored excitation and readout such as Arterial Spin Labelling (ASL), a technique capable of measuring cerebral blood flow (CBF) and perfusion in vivo.

Arterial Spin Labelling

In ASL cerebral perfusion maps are obtained without administration of contrast agents. Flowing blood is magnetically tagged at the carotid arteries with RF pulses. In particular, the spins of the flowing blood are inverted via special RF pulses (e.g. adiabatic or pulse trains) so that the relaxation is long and hence differ from the control as much as possible. After a short time, Post Labelling Delay (PLD), based on the blood bolus an EPI is taken in the region of interest (RoI) in the brain. The

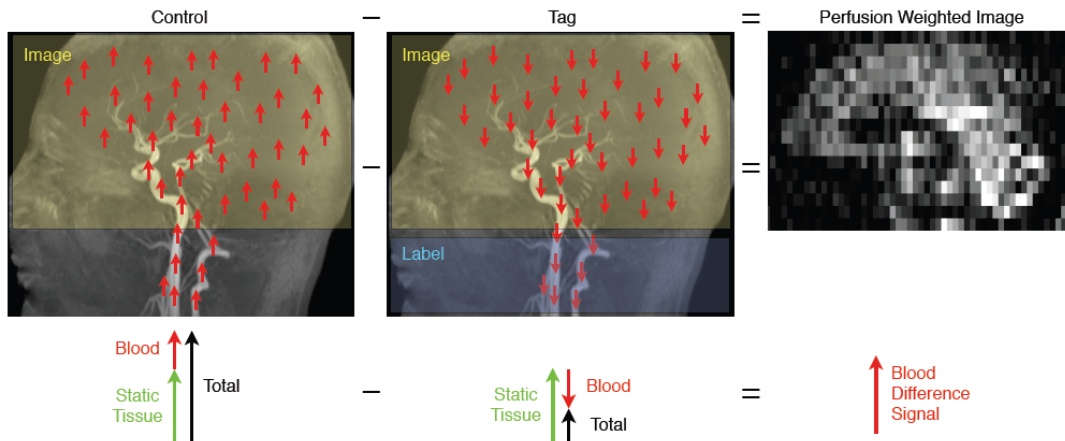


Figure 3.16: Arterial Spin Labelling process illustrated. In the control image the static and blood signals add up. When the blood is tagged (inverted), its signal is altered but the static signal remains uninverted. The subtraction of these two images will eliminate the static tissue contribution and provide only perfusion information. Reproduced with permission from [87].

difference between EPI images with (tag) and without (control) labelling is used to eliminate the static tissue signal and a perfusion map can be generated, see figure 3.16. The level of signal drop in other areas of the brain is assessed and information about its perfusion and flow can be extracted. It was first proposed in 1992 [53], and has since been under development and improvement.

ASL suffers from several problems and currently is mainly used only in research. The most prominent among these problems is the very low SNR of the method. The signal difference between the labelled and control images is very small even if the tagging efficiency is nearly 100%, i.e. completely inverted spins. Many signal averages are necessary to overcome this, which extends the imaging time and adds motion errors.

BOLD

The BOLD (Blood Oxygenation Level Dependent) effect was observed at the start of the 1990s in animal-based experiments [88, 89, 90]. It is known that oxyhaemoglobin is diamagnetic (i.e. essentially nonmagnetic) and that deoxyhaemoglobin is para-

magnetic. This means that deoxygenated blood has a shorter T_2^* and hence lower MR signal than fully oxygenated blood. During an fMRI experiment, the blood concentration is heightened due to neuronal activity as a result of the requested task. The oxygen consumption does not follow in proportion and that leads to a signal increase from the higher oxyhaemoglobin concentration. Figure 3.17 shows how fMRI data can be handled and presented to generate interesting insights of the brain.

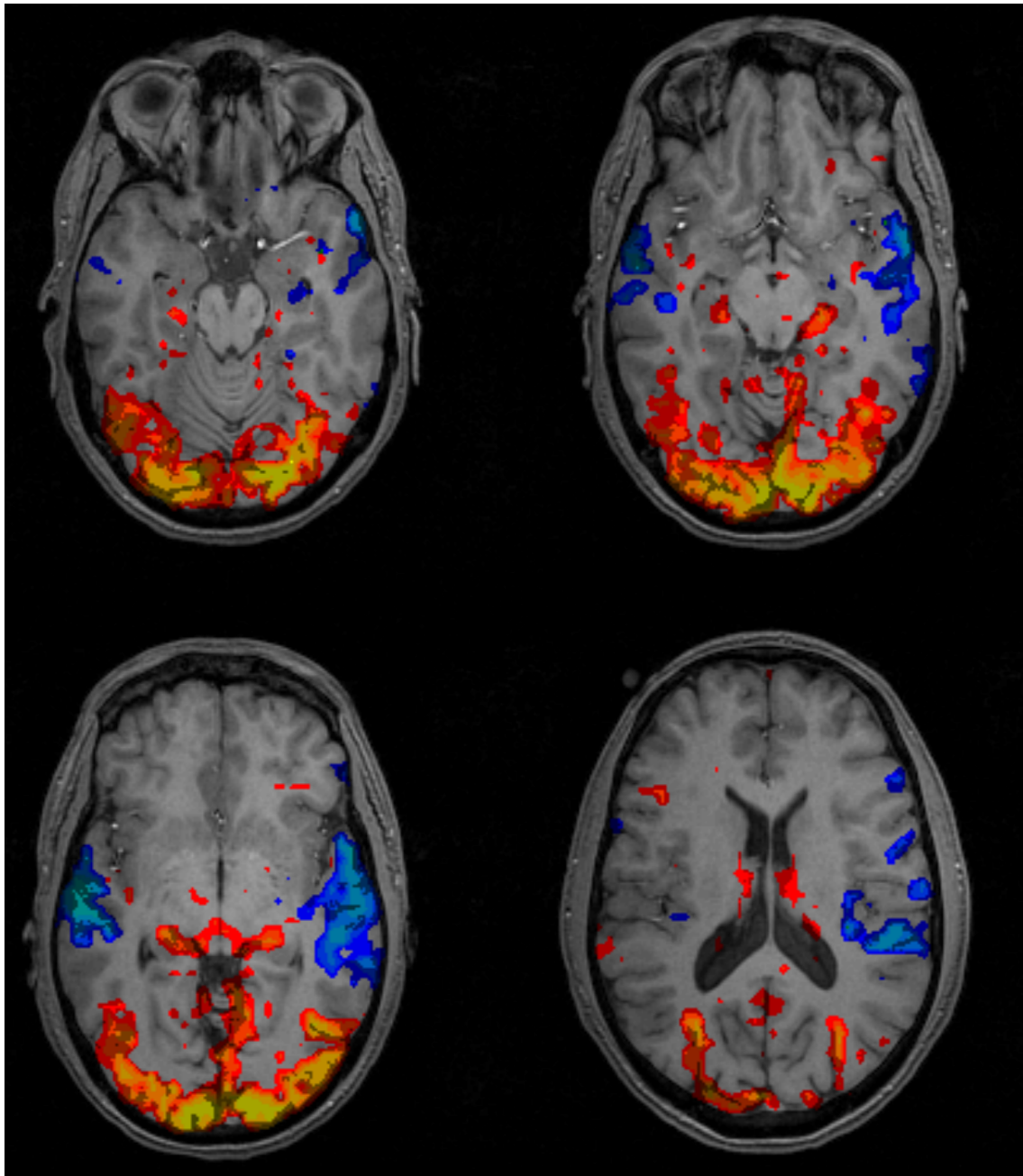


Figure 3.17: Functional MRI example with activation highlighted in blue, red and yellow depending on the BOLD signal strength. Simultaneous audio and visual stimulation was applied. The temporal MR signal (i.e. from the quick succession of images taken with EPI) from each area of the brain is put against the action potential response and based on how well it is correlated with it the area is considered active or inactive. Reproduced with permission from [91].

3.5 Instrumentation

3.5.1 The Magnet

Static high field superconducting solenoid magnets are used in MRI nowadays because they can generate very strong fields and have excellent field stability. Typical strengths in clinical scanners are 1.5 or 3 T but human magnets as high as 11.7 T have been constructed for human imaging (Neurospin Saclay, France) and 34 Tesla for spectroscopy (National High Magnetic Field Laboratory, FL). It is more challenging to achieve strong static field strengths as bore diameter increases due to requirements in more coil windings, space, power consumption and structural considerations due to the weight of the system. Consequently, the highest field strengths have been achieved in small animal imaging or MRS magnets.

The main magnet consists of thousands of metres of niobium-titanium windings or other specialized alloys immersed in liquid helium. The cooling in a modern scanner is done using liquid helium and until recently, liquid nitrogen, see figure 3.18. At these temperatures certain wires become superconductors, able to bear extremely high direct currents inducing a magnetic field [92]. The superconducting property allows high current to circulate a closed loop indefinitely in theory as long as fresh coolant is introduced and keeps the temperature low enough.

3.5.2 Shim Coils

The homogeneity of the main magnet is important for reliable MRI and becomes more crucial for higher field strengths. A uniform distribution translates to equal signal weight across the image, reduced susceptibility effects and in general a better T_2^* profile, figure 3.9. The magnet construction is of very high precision but local field inhomogeneities are present mostly due to the sample itself in areas such as transitions between tissues and air (e.g. sinuses). In order to correct these field disturbances additional shim coils are wound inside the main coil in order to generate

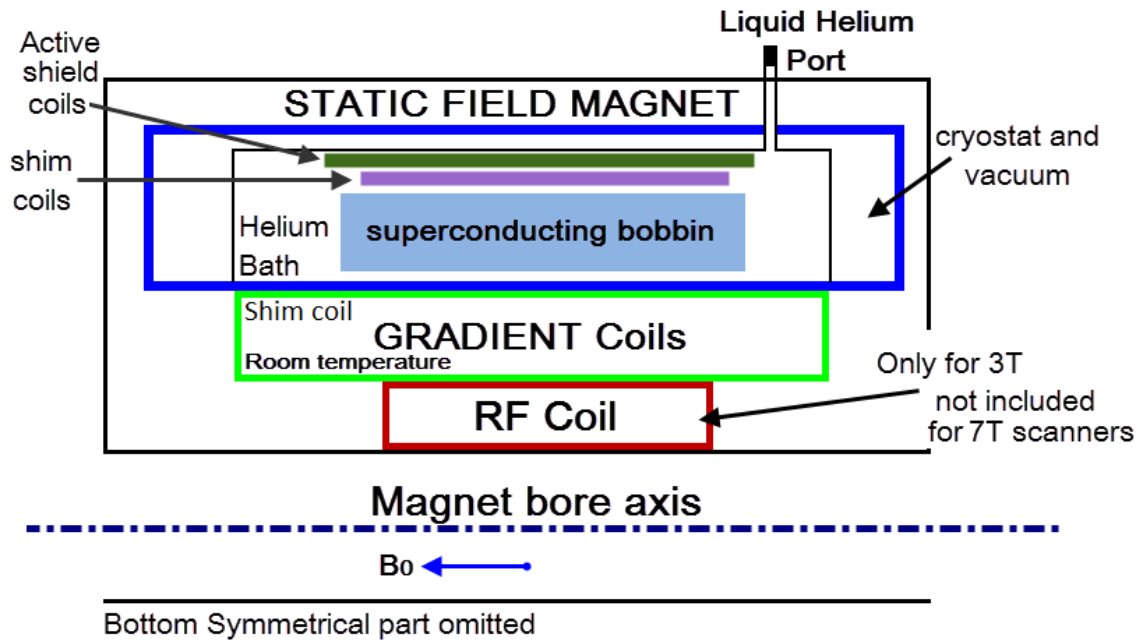


Figure 3.18: Schematic representation of liquid nitrogen and helium pump, superconducting coils and gradient system.

fields that either cancel or enhance the main one and optimize its homogeneity, figure 3.18. These are resistive coils capable of producing field corrections of several orders of spherical harmonics. During an MRI scan static $\mathbf{B}_0$ inhomogeneities can be mapped and eventually used to correct the final image.

$\mathbf{B}_0$ shimming is typically one of the first tasks that the operator has to perform before any scan. Field homogeneity is measured by running two scans with different TE and observing the phase difference. Shim currents are then adjusted by either the operator or more usually automatically to produce a homogeneous $\mathbf{B}_0$ field.

3.5.3 The Gradient Coil System

The gradient coil system comprises three resistive coils which can generate a linear static field variation in B_z along any direction in space. Gradient fields are used in conjunction with the RF coil for the image acquisition and are also positioned in the inside of the static field coil, figure 3.19. During a scan the coils are driven by fast switching strong currents. The switching frequency is in the audio range

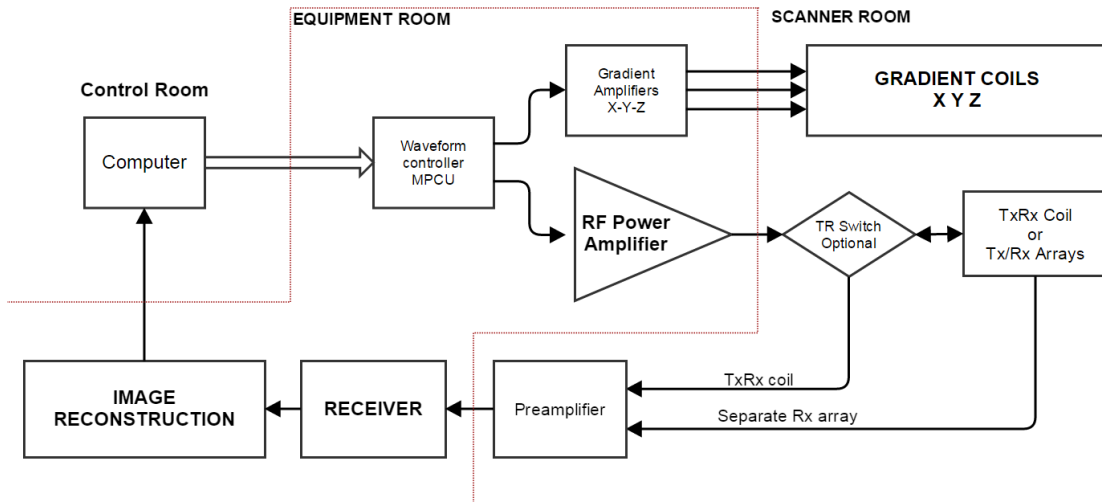


Figure 3.19: MRI scanner schematic diagram of main components. The sequence is loaded in the computer and fed to the waveform controller. The RF component is directed to the power amplifier and the gradient waveforms to the gradient amplifiers. Through the TR switch the pulse will be directed to the probe. The receive signal is amplified by the preamplifier and then fed to the receiver.

and Lorentz forces on the gradient windings oscillate against the metal former with audible frequencies producing the characteristic noise heard during MRI scans. The performance of the coils can be as high as 70mT/m for whole body human imaging but their performance depends on the switching speed. However, there is a physical and safety limitation to the switching speed due to the possible generation of eddy currents in patient and consequent nerve stimulation.

3.5.4 The Radio Frequency Coil

The RF coil is used to excite the spins of an area or a slice as well as to receive the useful signal. RF coils, as their name implies, function in the Radio Frequency range, transmit pulses and capture the raw signal for image formation. RF coils are designed to generate a rotational field, abbreviated B_1 , of the same frequency as the precessing spins. Either the same coil can be used for transmission and reception, or separate ones, and in clinical MRI the second option is used with localised receive probes. Current clinical scanners come with an embedded volume body coil that is

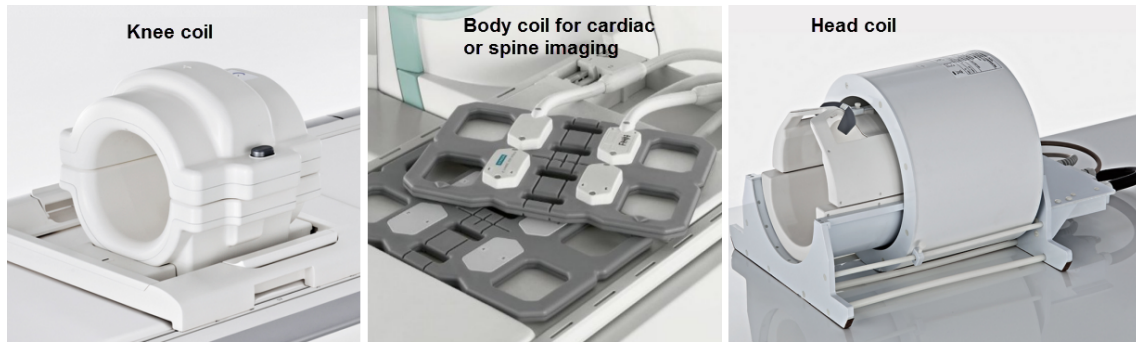


Figure 3.20: Typical clinical coils used for MRI at 3T.

placed just at the outside of the magnet inner bore. The patient then enters the scanner wearing the appropriate receive coil that looks like a helmet for a brain scan, flat pad for torso or cylindrical tubes for extremities, figure 3.20.

Two important parameters of an RF coil are the $\mathbf{B}_1$ field distribution and the sensitivity, and their combination determines the SNR. It is highly desirable to be able to excite all regions of a slice with the same 'strength' or flip angle since the latter depends only on the $\mathbf{B}_1$ field, equation (3.6). Uneven excitation will generate contrast within the FoV that does not stem from the proton density variation and hence can make diagnosis challenging or unreliable.

The effects of $\mathbf{B}_1$ inhomogeneity become stronger at higher fields due to the smaller wavelength of the Larmor frequency. At 3T for proton nuclei the value of f_0 is around 127 MHz and such a frequency in the human body has a wavelength of roughly 30 to 50 cm which is longer than the typical dimensions of an RoI. However, at 7 T the Larmor frequency for proton imaging rises to 300 MHz and the wavelength is about 15 cm. Such wavelengths are comparable to the body dimensions and the propagation of the wave is visible by creating maxima and minima within the body, figure 3.21. This will cause a variation in the flip angle and degrade the quality and reliability of the image. In some cases, field variation becomes so severe that the signal can be totally lost from some areas (drop-out). Thus, coil design methods for high fields concentrate on mitigating these effects with techniques such as RF shimming or parallel transmission that will be discussed later.

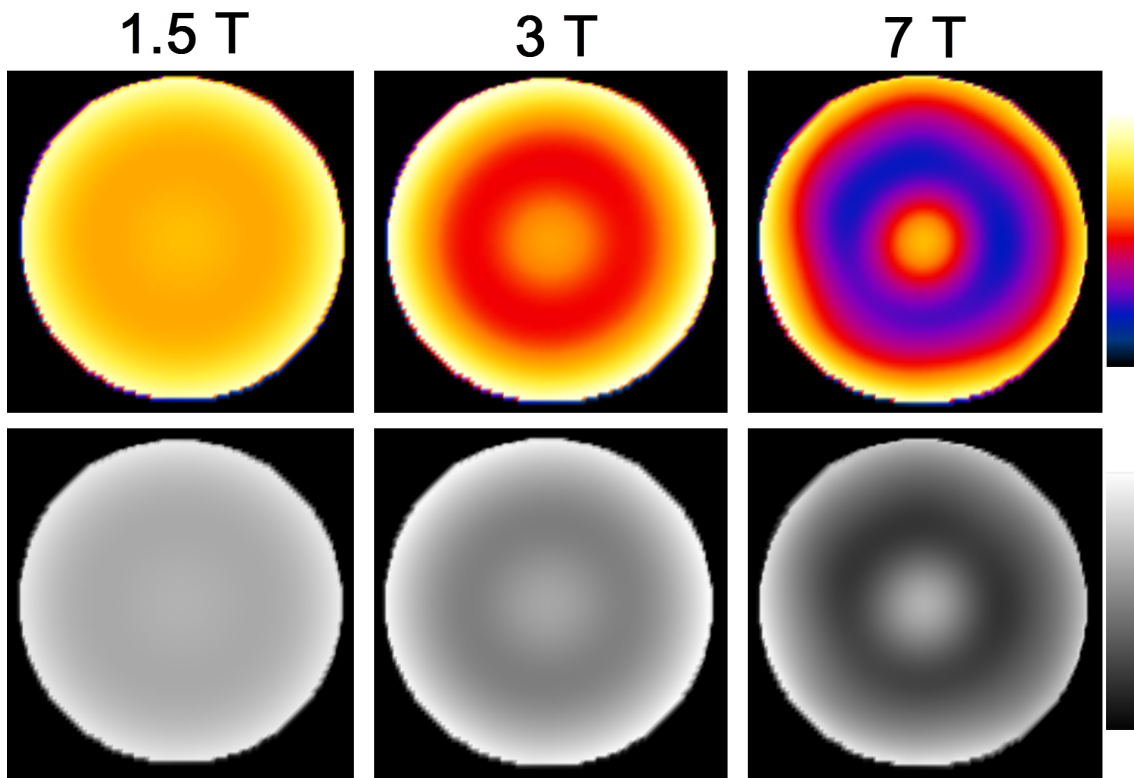


Figure 3.21: Distribution of the $\mathbf{B}_1^+$ field for a volume coil (left) and the equivalent brain scan slice for different field strengths. Reproduced from [93].

Another important factor of RF coils is their sensitivity; it is one of the main factors determining SNR which, as discussed earlier, poses a limit on how much of the high spatial frequencies in k -space will be covered. The latter holds the high resolution information of the image and hence high SNR means high resolution and fine detail visible, a feature highly desirable in medical imaging. It can also mean faster MRI examinations with fewer averages and reduced time in the bore for the patient; less power transmission and less RF heating. Exogenous means to improve SNR are an increase in the static $\mathbf{B}_0$ field or a decrease in the sample temperature. The latter is impossible for biological tissues so most effort has gone into constructing higher field scanners and optimising RF coil design and performance.

Two basic categories of RF coils exist in MRI; one is the volume coils that surround the ROI and the other includes the surface coils that are positioned over the ROI. There are two quantities that describe these characteristics, the coil volume,

V_c , and the filling factor, η . Equation 3.12 shows that both factors contribute to the SNR improvement but they cannot be optimised at the same time. Thus, volume coils have a high V_c but lack a high filling factor whereas the opposite is true for the surface coils. Usually volume coils provide a more homogeneous $\mathbf{B}_1$ with images from a large FOV but surface coils are positioned closer to the ROI and a better SNR can be achieved locally.

Abraham proposed a formula for the coil sensitivity [94]:

$$SNR \propto \eta \frac{\omega_0 V_c}{T_c \Delta f^{1/2}} Q \quad (3.12)$$

where η is the filling factor which is higher when the shape of the coil is closer to the examined object; Q is the quality factor; V_c is the coil volume; ω the Larmor frequency; T_c the coil temperature and Δf is the bandwidth of the receiver.

An RF coil can vary from a simple structure to a fairly complex design where multiple modes have to be taken in account; the coupling of different elements within it and the environment, load characteristics etc. These will all be discussed in detail in the next chapter. In the next section a brief description of the RF coil development timeline is provided.

3.6 Development of the Radio Frequency Coil

Simple solenoid coils were used for the first NMR experiments and were placed transversely relative to the static field direction in order to tip the spins. Solenoid coils are axial resonators, i.e. their magnetic field is parallel to their axis, and hence their geometry and construction is simple and their performance reliable. They can be fabricated by one or multiple windings of wire or a by thin copper foil wrapped around a cylinder forming the loop gap resonator.

The need for resonators with a cylindrical geometry that is co-axial with the magnet for better functionality, brought saddle shaped coils, figure 3.22, in which

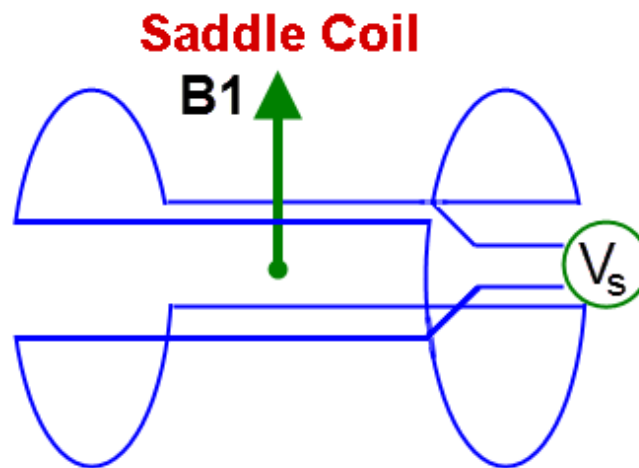


Figure 3.22: Saddle shaped coil schematic.

the $\mathbf{B}_1$ field is perpendicular with respect to its main axis as a solenoid. As the main field strength increased the Larmor wavelength became very close to the saddle coil conductor length and hence these coils became impractical for MRI. The introduction of slotted-tube resonators was the means to overcome the previous limitation. It had distributed capacitance and lower inductance and hence it was possible to operate at higher frequencies. This coil was the predecessor of the Alderman Grant coil [95] in which guard rings were introduced in order to lower electric fields, decrease RF heating of samples and increase the SNR.

The surface coil was introduced in 1980, [67] and was a simple loop coil. This design provided another option to MR scientists, since until then the RoI was fully enclosed by bulky volume coils. The surface coil is closer to the ROI and can achieve better SNR from that area but it has more restricted sensitivity in other areas and a very inhomogeneous $\mathbf{B}_1$ field.

A significant improvement in surface coil function was the use of more than one coil around the sample in order to improve reception and hence sensitivity and SNR [96, 97, 98]. The so-called Quadrature drive [99] allows better field homogeneity but also improves the sensitivity of the coil. In quadrature drive the probe has two feeding ports which are excited by currents with a 90° degree phase difference.

Quadrature driving enables a $\sqrt{2}$ increase in SNR with an associated need for less RF power.

3.6.1 Phased Array

A further improvement in B_1 distribution was achieved by using more than two surface coils driven in phase, or quadrature drive [100, 101, 102, 103]. Another head coil consists of multiple surface coils forming a dome structure [104], having very good sensitivity for localised ROI in the brain. Zhang and Wu [105, 106, 107] have developed microstrip surface coils based on traditional transmission line theory and multi-channel implementations were made by Adriany et al. [108, 109].

The phased array coil introduced in 1990 by Roemer [110] was the ground work for the receive arrays used today. The principle of placing multiple loops in any desired shape and geometry was described with methods to minimise the coupling between the elements. Distinct sensitivity profiles are desirable for parallel imaging where a reduced number of k -space lines are acquired; the missing lines are filled retrospectively from the spatial profiles of the elements. Coupling between the elements implies noise for parallel imaging. The most prominent techniques are SMASH [98], SENSE [111] and GRAPPA [112]. Phased array coils and parallel imaging techniques are the predecessors of parallel transmission methods.

3.6.2 The Birdcage Coil

A milestone in RF coil design for MRI was the development of the Birdcage coil [113] which allowed high quality whole body imaging. The resonator consists of two rings of the same diameter placed concentrically and in parallel and multiple rungs connecting them. The rungs, also called legs, are driven by currents in phase in such a way that the current in the first and the last rung form a cosine, figure 3.23 and equation (3.13). This structure uses the principle of standing waves and forms a B_1 field perpendicular to its axis. A birdcage coil has an excellent B_1 distribution

profile and that is why it became the mainstream coil model used until high field imaging was introduced.

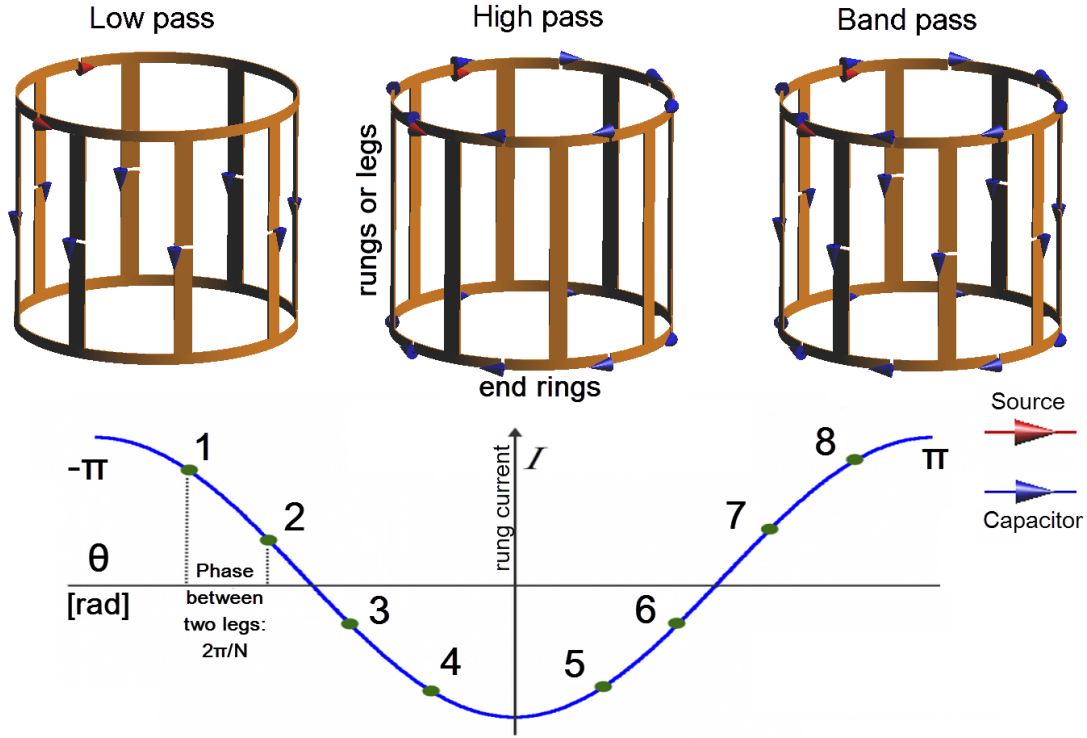


Figure 3.23: Birdcage coil design types (top), current for each rung has a phase and amplitude difference with the others so that the current profile forms a cosine (bottom).

For a Birdcage with $2N$ rungs:

$$I_n = I(2N)\cos(\theta_n + \phi) \quad (3.13)$$

for $n=0, 2N-1$, where I_n is the current that flows in the n^{th} rung, θ_n is the relative angle position of the rung and ϕ is the spatial phase factor which is defined by the driving mode of the birdcage.

As expected there have been many developments in birdcage coil design. Some of these are the high-pass birdcage [114] with capacitors on the rings, a half birdcage design in order to gain better access to body parts such as torso or limbs [115], an additional conduction ground sheet for $\mathbf{B}_1$ homogeneity improvement [86] and elliptical shapes to increase filling factor and SNR [116, 117, 118]. The hybrid

birdcage coil, where high and low pass parts are combined [119], functions as a band-pass filter and was a further advancement. In an attempt to improve $\mathbf{B}_1$ by using the birdcage coil principle and at the same time to avoid its complex design the Boligner coil was proposed [120].

The transverse electromagnetic (TEM) resonator, which is another variant of the bird-cage coil, was described in 1993 [121] and further developed [122] from the same group. The TEM resonator was introduced in order to be used in high field imaging and particularly at 4 T and above. In a TEM resonator, the lumped capacitances are replaced by transmission lines and tuning is done via copper stubs.

3.6.3 Other Coil Concepts

The Litz coil, used mainly for small bore diameter and high field applications, was a useful addition for small animal imaging [123]. The coil conductor is knitted in such way that the current distribution is near to ideal and hence it has an excellent $\mathbf{B}_1$ field distribution. However, it has the same problem as the saddle shaped coil in whole body applications. Scaling up to larger bore diameters or stronger magnetic fields the overall conductor length is comparable with the wavelength and hence the Litz design becomes impractical.

Recently, Brunner and Pruessmann [124] suggested another departure from the conventional MR coil. RF probes are substituted by patch antennas on the edges of the bore magnet. The method, known as travelling wave excitation, exploits the constructive interference of plane waves. The concept is promising and some results can be found in Andreychenko's and Mallow's studies [125, 126].

The concept of Parallel Transmission (pTx) was proposed by Katscher in 2003 [127] in order to mitigate the $\mathbf{B}_1^+$ inhomogeneity at ultra high field scanners. In pTx, an array of surface coils is used, similar to the phased array, and each element transmits with different amplitude and phase relative to the reference wave. The pre-calculated constructive or destructive interference of the electromagnetic waves

in the FoV help to improve the field profile the same way as shim coils homogenise the $\mathbf{B}_0$.

There are two ways to approach parallel transmission, RF shimming described by Mao et al. in 2006 [128] where each RF pulse is maintaining its sinc profile and only the phase and amplitude are altered. The second method is an independent transmission from each of the transmit channels where the waveform of each RF pulse can vary in order to generate an excitation profile. An implementation of the method was described by Setsompop in 2008 [129, 130].

Since its introduction at the beginning of the previous decade, pTx has been the subject of extensive research and development. Many pTx power amplifiers have been distributed in the MR community and interesting results have arisen from the use of pTx methods. Some studies with interesting results can be found in [131, 132, 133, 134, 135, 136, 137].

3.7 Conclusion

In this chapter, an introduction to Nuclear Magnetic Resonance properties of the matter has been provided. In particular, the means by which some nuclei are magnetically active and align with external magnetic static fields has been described. The basic NMR experiment, FID, has been described as well as the main concepts behind NMR such as excitation, relaxation etc.

MR Imaging basic technique were described including contrast manipulation, slice selection and spatial encoding methods. A reference has been made to other components such as the sequences and the receive chain. Finally, a thorough description of the MR instrumentation was given with particular emphasis on the RF coils, their development timeline, and some important and novel designs.

In the next chapter, a detailed description of RF coil theory, design considerations and construction methods will be provided. An attempt will also be made to

construct a transmit coil with a separate receive array to operate in an ultra high field 7 Tesla scanner for vessel wall imaging.

Chapter 4

Radio Frequency Coil Design Considerations

4.1 Introduction

Any scientific experiment requires a suitable ‘mediator’ between the physical phenomenon and the output that is used for assessment. The ideal measuring tool would have the least effect on the examined subject but with high sensitivity to its variations. When referring to the Nuclear Magnetic Resonance (NMR) experiment this mediator is the RF coil as introduced in the previous chapter.

In particular, the examined phenomenon in NMR is the intrinsic property of certain nuclei to precess at a particular Larmor frequency when subjected to a strong external static magnetic field. If the nucleus is magnetically active (has a net spin) then the elemental magnetisation aligns with the static field. The RF coil needs to be sensitive to detecting the precessing transverse component of the magnetisation and even if there is an abundance of nuclei in a sample, the signal is very weak and the task is not a trivial one. Figure 4.1 shows the static field, $\mathbf{B}_0$, and the magnetisation, M , being aligned and precessing around it as well as the electromotive force induced in a coil placed appropriately.

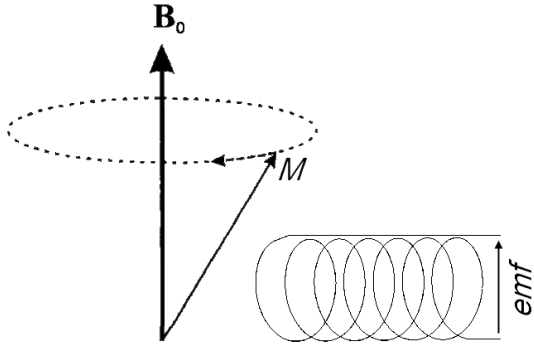


Figure 4.1: The M vector is the magnetization of the nucleus due to the intrinsic spin and its circular motion is the Larmor frequency precession approximation. When closed conductive loops are near alternating magnetic flux, such as the one generated by M , an electromotive force, emf , is induced generating a current to oppose it.

The total magnetisation from N nuclei can be derived via equation (4.1). Few parameters can be manipulated in order to strengthen the magnetisation when it comes to imaging biological tissues. The temperature and the number of protons have fixed values and in practice the static field is the only parameter that can boost the signal. The value of M is proportional to the $\mathbf{B}_0$ strength but the increase of the latter has implications on the installation cost and management:

$$\mathbf{M} = N \frac{\gamma^2 s(s+1) \hbar}{3k_B T} \mathbf{B}_0 \quad (4.1)$$

where N is the number of nuclei present in the sample, γ is the gyromagnetic ratio, s is the nuclear spin, $1/2$ for Hydrogen, $\hbar$ is Planck's constant over 2π , k_B is Boltzmann's constant and T is the temperature of the sample.

Since $\mathbf{B}_0$ is fixed, optimisation of the receiving RF probes is critical. However, transmit and receive coils contribute to the quality of the image in different ways and this will be discussed in this chapter. Even though not explicitly referring to RF safety here, it is necessary to build a device that: a) generates the required $\mathbf{B}_1$ field for the application; and b) is safe to be used for human imaging. The theory of radio frequency coils will be discussed along with an attempt to construct and electromagnetically characterise a coil for imaging the carotid arteries at 7 Tesla.

4.2 RF Coil Design Theory

A coil consists of a conductor with capacitance, C , inductance, L , and resistance, R ; the latter is introduced by the loop and the component equivalent series resistances (ESR). The simplest case of an RLC circuit comprises a circular loop and resonates at a single frequency dictated by the condition $LC\omega^2 = 1$. This occurs when the reactance of the inductor $j\omega L$ is equal to the reactance of the capacitor $1/j\omega C$. The coil can be treated as a load in a transmission line and the two main aims are to have a resonance at the Larmor frequency, and to optimise the power transmission from the source.

The fundamental theory needed for the description of a resonant coil as a load in a transmission line can be found in the literature. In the next subsections, tuning; matching; balancing as well as use of parallel transmission will be described.

4.2.1 Tuning and Matching

In order to achieve 100% power transmission from the source to the load the reflection coefficient, Γ , must be equal to zero. Solving for $\Gamma = 0$ in equation (4.2) the condition for an optimised match can be derived as shown. In fact, the amplifier/receiver impedance, Z_S , and the characteristic impedance of the transmission line, Z_0 , are set at 50Ω based on conventions in RF engineering. However, the coil impedance exhibited by the coil, Z_L , is not 50Ω and needs to be adjusted.

$$\begin{aligned}\Gamma = 0 &\Rightarrow \frac{Z_L - Z_S}{Z_L + Z_S} = 0 \\ &\Rightarrow Z_L = Z_S \\ Z_S = Z_L &= Z_0 = 50\Omega\end{aligned}\tag{4.2}$$

The RF probe as any circuit with passive components demonstrates a complex

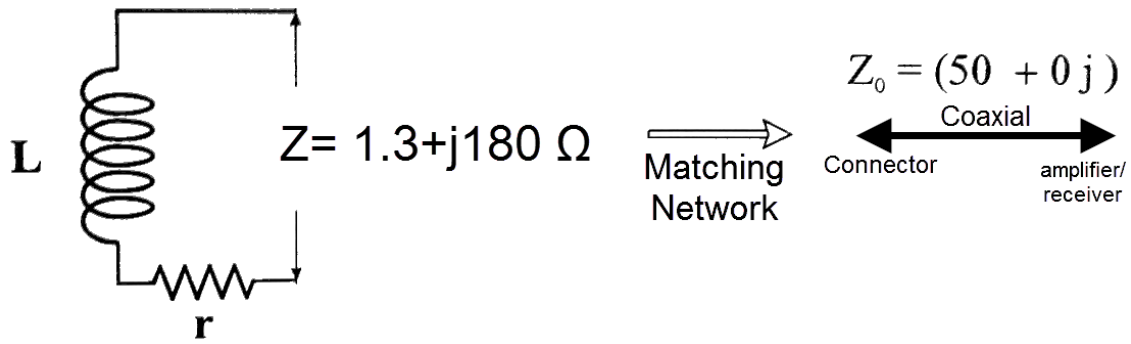


Figure 4.2: Schematic showing the need for an impedance transformation network that would take the mostly inductive impedance of an un-tuned loop to the matched condition at the Larmor frequency.

impedance $R + jX$ where the real part, R , represents the total wire resistance, usually a small number, and the imaginary part is the reactance of the inductance, see figure 4.2. The impedance is far from the purely resistive 50Ω and hence a ‘matching network’ is introduced between the coil and the coaxial cable. The matching network is a two-port input-output device comprising passive components and transforms the impedance of the coil to 50Ω so that the matching condition is met.

An optimised power transmission, however, is not the only condition for a functioning coil. The coil needs to operate at the Larmor frequency in order to resonate with the precessing spins. The tuning is achieved by adding capacitance accordingly in order to meet the resonance condition $LC\omega^2 = 1$. The impedance of a resonating circuit, if measured with a spectrometer, exhibits a very large resistive part with almost no reactance. In particular, the real and imaginary parts of a resonant circuit over a frequency range show a response as shown in figure 4.3, along with a description of the matching process.

It is interesting to note here that a resonant loop exhibits a very high impedance attributed to the fact that the current is ‘trapped’ oscillating from one form of energy to the other. The phenomenon of an apparent high impedance of a resonant circuit is utilised in decoupling methods. Currents in a primary loop are minimised by introducing a secondary resonant loop in series with it. Detuning methods based on this technique will be discussed later in this chapter as well as utilised for the

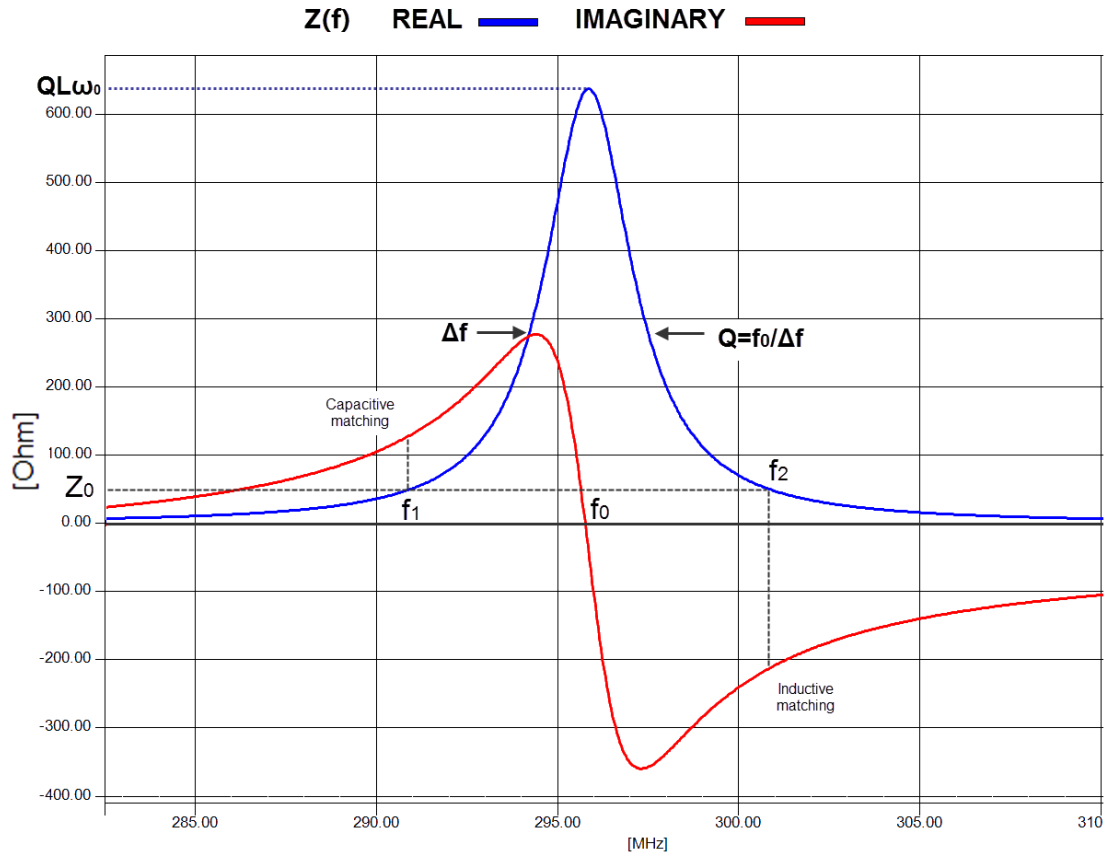


Figure 4.3: Real and Imaginary components of the complex impedance of a resonant loop. The resistive part becomes very large with the imaginary approaching zero. For matching to be achieved the coil needs to slightly deviate from its tuning frequency so that f_1 or $f_2 = f_0$. There, the real part is equal to Z_0 and the remaining reactive component can be compensated by an opposite value passive component. The most generic case is when capacitive matching is used and hence the f_1 is selected to be the f_0 . A capacitor then can be added in series and bring the total impedance seen to $50 + 0j\Omega$.

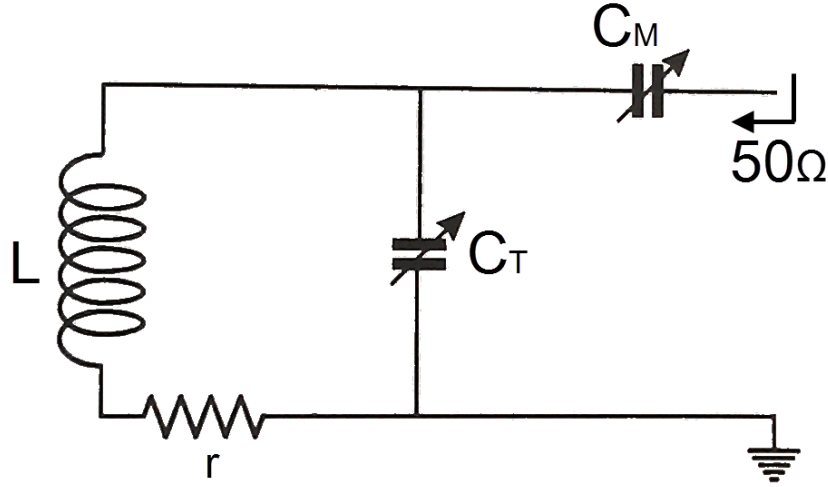


Figure 4.4: Series capacitive matching network with two variable capacitors, C_T and C_M for tuning and matching, respectively. The process is almost never analytical and hence variable capacitors are used that are turned in order to achieve the required matching by monitoring the reflection with a network analyser.

device under consideration.

Referring to figure 4.3 again, 50Ω real resistance can be achieved at one higher and one lower frequency compared to f_0 . For matching and tuning a combined methodology is followed where the main resonance is shifted so that either f_1 or f_2 is the Larmor frequency. In either of these two positions the real part of the impedance is 50Ω and the imaginary part can be either positive or negative with the most common option to be f_1 . The positive reactance is then nullified by adding in series a component with negative reactance, i.e. a capacitor. Figure 4.4 presents the generic case of capacitive tuning and matching with the use of variable capacitors.

The process described here is series capacitive matching and this is the most conventional method for matching coils to 50Ω . This is due to the suitability of the available variable capacitors that have a sufficient range, tolerance and accuracy to optimise the process. Other methods include inductive matching, i.e. using variable inductors, matching using transmission lines (stubs), broadband matching (communications), active matching and Balun matching and the reader can refer to [138] for more information.

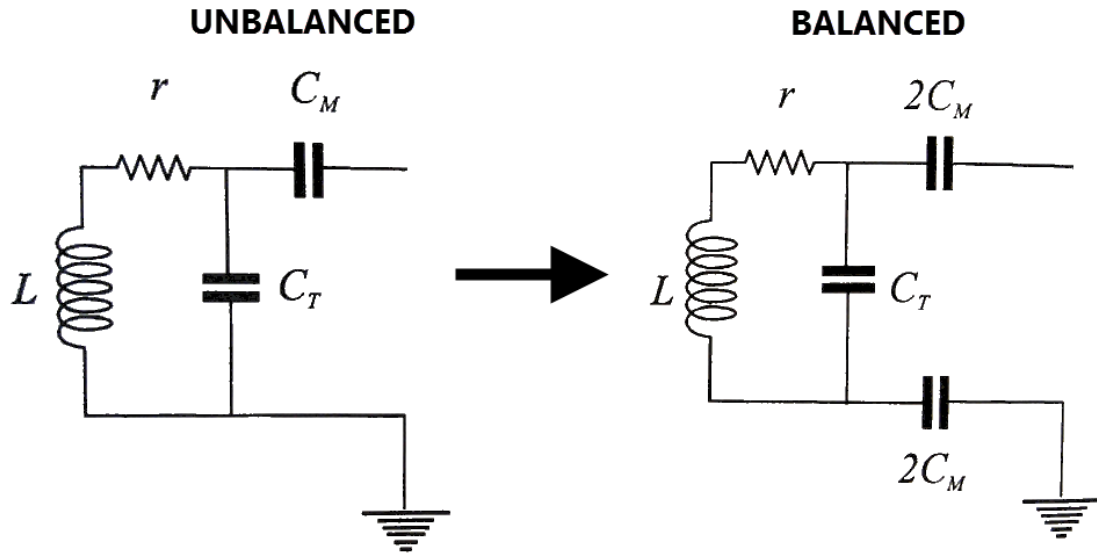


Figure 4.5: A standard unbalanced circuit is shown on the left where the shield kept at the ground. On the right, the symmetrised circuit is proposed for two $2C_M$ capacitors and the potential across C_T is halved with respect to ground. That means that the sample now experiences half the potential and the loss is a quarter relative to before since the deposited power is proportional to the square of the voltage.

4.2.2 Balancing the Probe

The time varying electric field associated with a resonating clinical coil causes dielectric and radiation losses in the sample which reduce the sensitivity of the coil. The aim for a coil builder is to minimise, if possible, the electric field in order to mitigate these effects [139]. One solution is electrical balancing of the probe with respect to the ground as well as the use of a low inductance design. The basic principle of balancing is to reduce the two terminal potentials by constructing a symmetrical matching network. Figure 4.5 shows the most common solution where the matching capacitor C_M is replaced with two capacitors with double the value, $2C_M$, placed between the two terminals of the tuning capacitor and the coaxial cable. The currents, in the ideal case, are the same and opposite hence generating the same voltage difference as previously but only half of it with respect to the ground, i.e. from $\pm V \rightarrow 0$ to $+V/2 \rightarrow -V/2$.

During the coil loading, parasitic capacitances form between the metallic con-

ductors and the conductive tissues, see figure 4.6, with these capacitances being of the order of a few pF. Due to the electric properties of the tissues, see table 4.1, magnetic and dielectric losses are present. For the magnetic losses, i.e. eddy currents, very little can be done; but for the dielectric losses, i.e. the material's inherent dissipation of electromagnetic energy, which are proportional to the square of the voltage, a symmetric configuration as shown in figure 4.5 can reduce them by a factor of four.

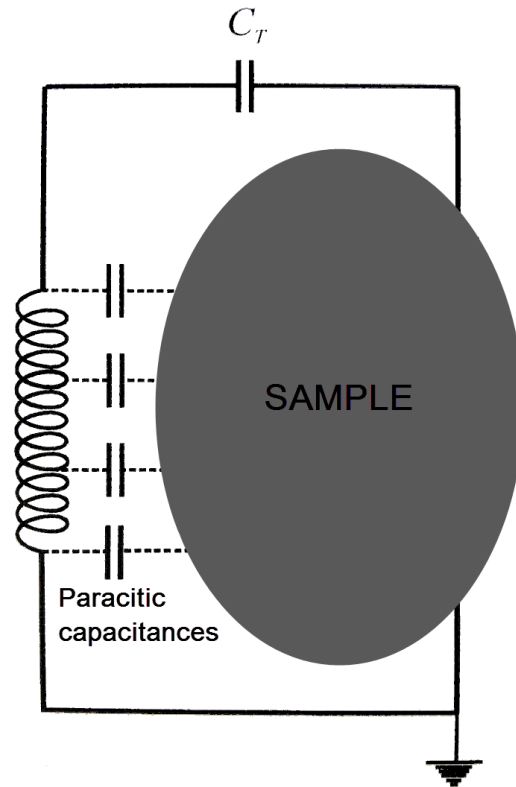
$\sigma/(\omega\varepsilon)$	Current conduction	Field propagation
0		perfect dielectric lossless medium
$\ll 1$	low-conductivity material poor conductor	low-loss medium good dielectric
≈ 1	lossy conducting material	lossy propagation medium
$\gg 1$	high-conductivity material good conductor	high-loss medium poor dielectric
∞		perfect conductor

Table 4.1: Human tissues have a relative permittivity (ε_r) from 15 to 65 and conductivity (σ) in the range of 0.2 to 1.5 S/m. It can be estimated that materials with these properties lie in the lossy conducting media group, exhibiting dielectric and conductive behaviour. Thus, tissues are poor quality dielectrics with resistive and dielectric losses. The ε value used here is the real part of the complex permittivity calculated as $\varepsilon = \varepsilon_0 \cdot \varepsilon_r$.

The second proposed methodology to reduce losses is the use of a low inductance design. In such a design the capacitance required for the resonance $LC\omega^2 = 1$ is bigger compared to a higher inductance loop. Large capacitors can store more electrical power under smaller voltage and this results in lower $\mathbf{E}$ -fields which is the desired effect. In addition, large capacitance creates stable coils when loaded since the ratio of the coil capacitance to the parasitic capacitance is higher, leading to a smaller resonance shift on loading or by insertion into the bore. However, high currents are associated with high mutual inductance in multiple loop coils and generate the undesirable coupling that will be discussed later.

Radiation losses depend on the wavelength with respect to the conductor dimensions. As a rule of thumb [138, 140], if a circuit is of the order of or bigger than one

Figure 4.6: When a subject is positioned in the vicinity of the coil the interaction that follows describes the coil loading. Parasitic capacitances form between the metallic conductors and the conductive tissues with the air in between acting as the dielectric. The same can happen with the metallic walls of the magnet bore. The effect is witnessed by a shift in the resonance frequency and a reduction of the Q-factor due to the additional losses from the leakage.



twentieth of the wavelength ($\lambda / 20$) then the radiation losses become significant. For a coil with such dimensions its parasitic capacitance increases and the quality factor (Q-factor) falls further. The distribution of the capacitance around the coil is a very common solution where the total required capacitance is divided into smaller capacitances in series and placed at equal distances on the loop. This process is called 'segmentation' and shortens the maximum wire length in the coil, reducing the radiation losses.

To summarise, for the reduction of losses due to the electric fields and hence the increase of the sensitivity (Q-factor), three principles have to be followed:

- Low inductance design.
- Maximum wire length less than $\lambda/20$ and use of segmentation.
- Symmetric matching configuration as shown above.

Another source of loss, however, is the common mode effect. Once the coil is made symmetrical, in the ideal case the currents and voltages are opposite and the

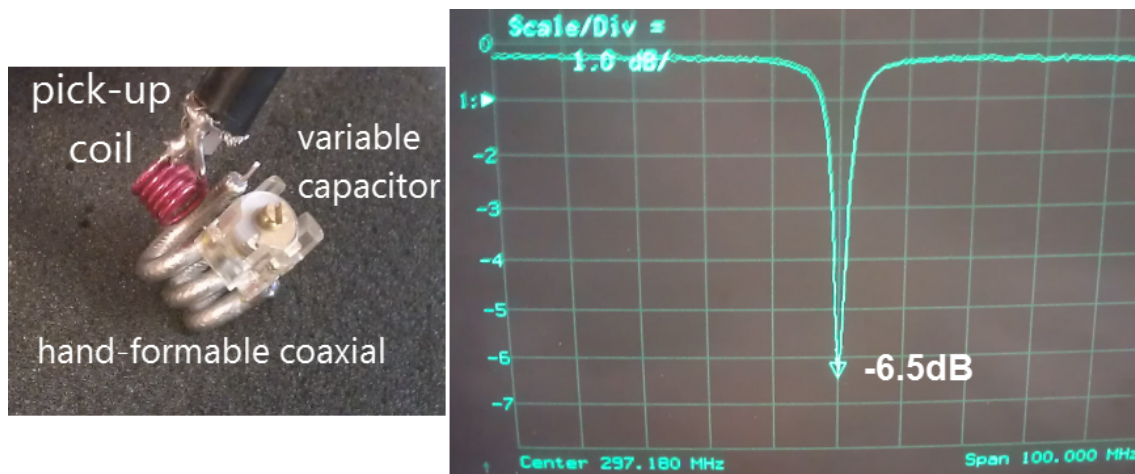


Figure 4.7: A very small diameter pick-up coil is used to assess the resonance of the shield trap before it is connected to the circuit. A variable capacitor is turned until the S_{11} resonates at the Larmor frequency, 297.18 MHz at 7 Tesla. Usually, the BALUN can be tuned in an adjacent frequency in order to reduce strong coupling with the main coil as well as shielding can be used for this purpose.

coil operates in ‘differential mode’. However, any coupling of the coil to the coaxial cable shield will generate currents that are not associated with the differential mode and that will cause a perturbation. This translates into additional loss since it affects the tuning and matching, produces a sensitivity to background noise and, for long enough cables, leads to radiation loss. Currents flowing in the outer shield can be hazardous to the subject since they can cause burns if the cables are touched.

A solution for the common mode effect is to eliminate the currents flowing in the outer shield by adding a high impedance in series. The most common method is the use of a shield trap used between the matching network and the coaxial cable. The latter, pictured in figure 4.7, is a piece of coaxial cable wound in a few turns with its shield connected to a capacitor; the inductance from the shield and the connected capacitance resonate at the Larmor frequency. This creates a high impedance that suppresses the currents and reduces the common mode effect. This is known under the term ‘shield trap’ or ‘BALUN’ that stands for BALanced-UNbalanced since they contribute to the overall stability of the circuit. For long coaxial cables multiple shield traps can be used to suppress the standing wave maxima.

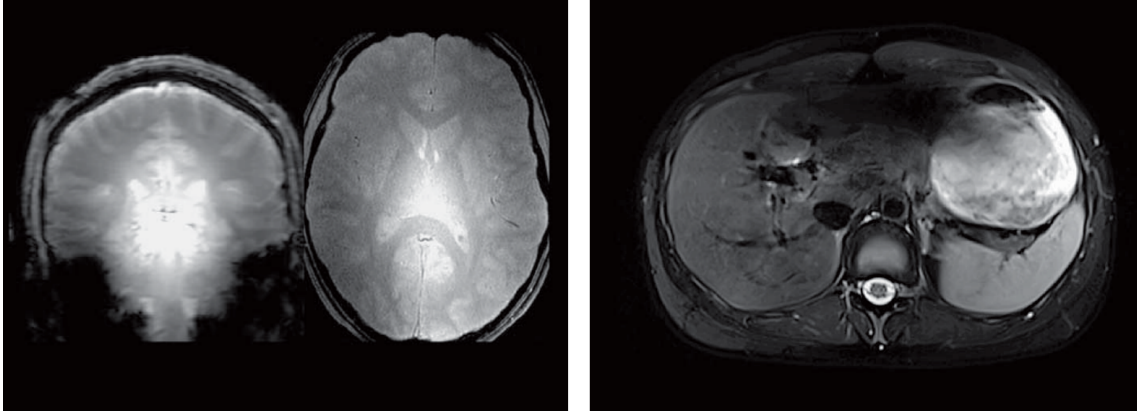


Figure 4.8: Flip angle inhomogeneity resulting from wavelength effects in the brain at 7T (central brightening) and liver at 3T (drop-out). Spatial variations in the transmit efficiency, and therefore the flip angle, are more problematic than the receive inhomogeneities since they lead not only to image shading, but more importantly, image contrast alterations. Reproduced from [142].

The subject of balancing and matching probes is a fairly wide and complex topic and more details can be found in [139, 141, 138, 140]. For the construction of the coil later in this chapter all the aforementioned principles and methods will be utilised in order to minimise losses and to increase the stability and sensitivity.

4.2.3 Parallel Transmission

At 7 Tesla the Larmor frequency is approximately $\omega_0 = \gamma \cdot \mathbf{B}_0 \simeq 300MHz$ and the wavelength is of the order of a few centimetres in human tissues, see equation (4.3). This has major implications in human imaging at ultra high fields since the field distribution varies significantly within the dimensions of the subject. Local maxima can be observed as bright areas and minima as dark areas where signal is lost, see figure 4.8.

$$\lambda = \frac{c_r}{f} \text{ for } c_r = \frac{1}{\sqrt{\varepsilon_0 \varepsilon_r \mu_0 \mu_r}} \quad (4.3)$$

For a typical tissue such as muscle, $\varepsilon_r = 58$ and hence the above equation gives $\lambda_{muscle} \simeq 13cm$.

The most promising method to tackle the inhomogeneity issue is to use parallel

transmission (pTx). The latter involves the use of multiple transmit elements in order to adjust for these $\mathbf{B}_1$ inhomogeneities. Each coil transmit profile is known and mathematically it can be calculated how each source will vary compared to others in order to create homogeneity within the RoI. The most straightforward solution is ‘RF shimming’ where the variables to change for each element are the RF magnitude and the phase. Thus, the RF pulse in its original form is transmitted from each element but small variations in the phase and magnitude lead to specific superposition and cancellation of waves and thus mitigate the inhomogeneities. Before and after images will be shown in the Chapter 6 where RF shimming is used.

4.3 MR Application

4.3.1 Black Blood Imaging - DANTE

Following up on the clinical background chapter, an application is now presented for high resolution carotid artery imaging. The aim is to design and construct a suitable RF coil for the assessment of the arterial wall plaque structure at the carotid artery bifurcation at ultra high field, i.e. 7 T. Before doing so, an overview of the effects that moving spins have on the MR signal will be provided as well as a discussion of how these effects are utilised in angiographic methods.

Flow Artefacts

Moving protons in blood vessels cause artefacts due to the replacement of the excited spins with new ones or by the accrual of unwanted phase during read-out, see figure 4.9 for an example of the ‘flow artefact’. The velocity induced phase of moving fluids generates ghost images and in particular, the pulsatile laminar flow within the vessels can produce a complex multilayered band that usually propagates outside the head in the PE direction, figure 4.9. The appearance of blood in SE and GE sequences also differs; in the first, blood generally appears dark and in the second

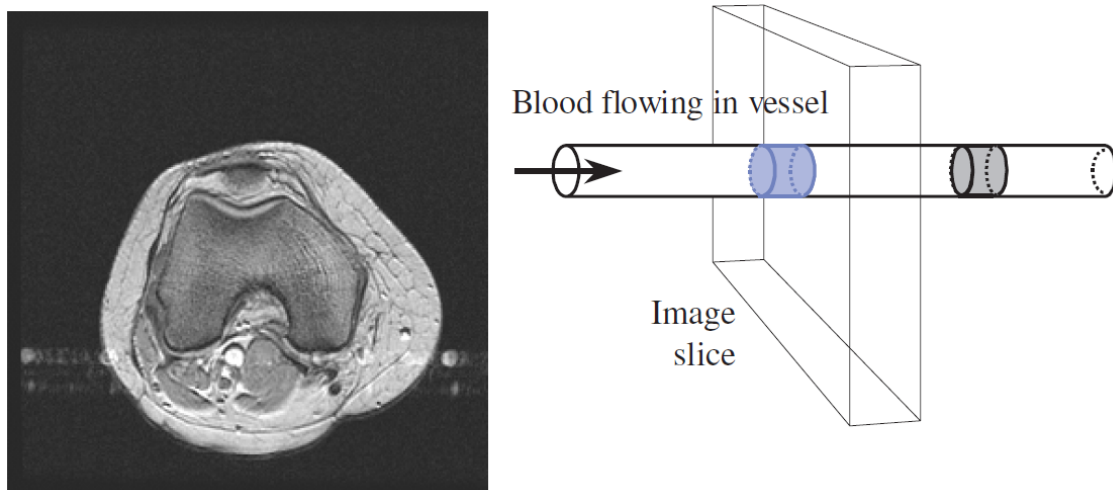


Figure 4.9: Laminar pulsatile flow can cause artefacts along the phase encoding direction as shown in the knee image on the left. The right image shows why this happens; in a spin echo sequence, the time between the 90° and 180° pulses allow blood to be replaced in the excitation slice which then fails to generate an echo. Reproduced from [79] with permission.

blood appears bright, see figure 4.10. For these reasons, vessel wall imaging cannot be performed without the use of specialised methods that mitigate these effects or in fact take advantage of them.

Although moving spins can degrade the signal, MR angiography (MRA) techniques take advantage of the flow effects in order to generate images of the circulatory system. The signal can be extracted by endogenous contrast of the flowing spins due to flow dependent changes in the longitudinal magnetisation, known as time-of-flight effect (TOF), or by flow dependent changes in the phase of the transverse magnetisation (phase contrast, PC). The vessels can be assessed either by using bright blood techniques where the blood is highlighted and the static tissue is suppressed or black blood (BB) imaging for cross sectional assessment of the vessel wall, figure 4.11.

Depending on the investigated disease, bright or black blood is prescribed but usually both provide unique and overlapping information for a thorough diagnosis. For the carotid arteries for instance, bright blood MRA can provide an overview of stenoses along the arteries but the atherosclerotic plaque can only be assessed by black blood imaging techniques. Prominent methods for BB imaging are double

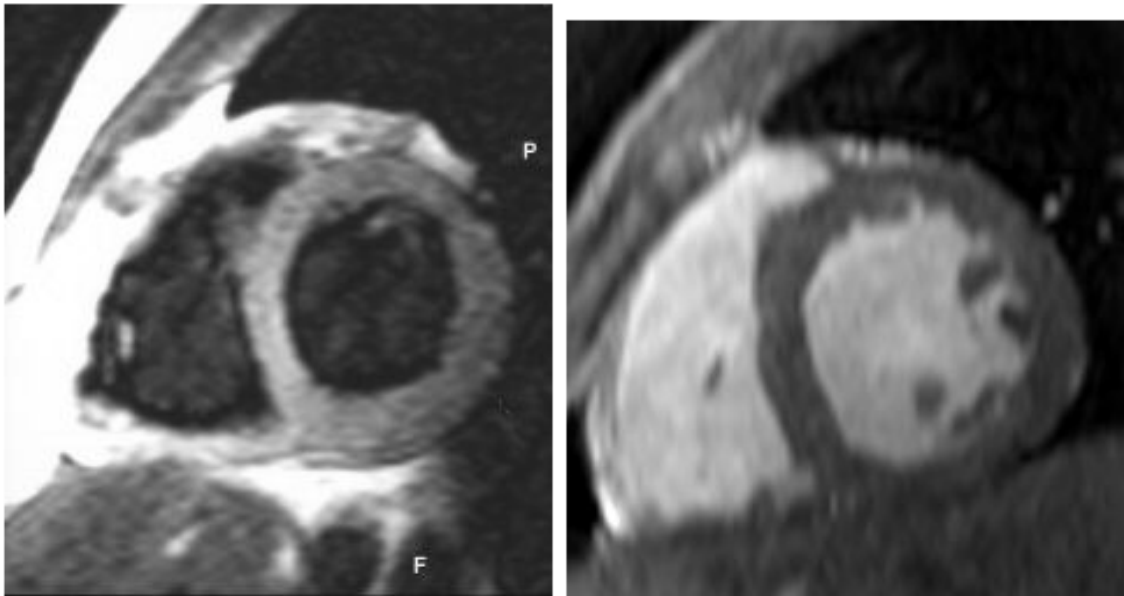


Figure 4.10: Black-blood imaging (left) using a spin echo sequence and bright blood imaging (right) when a gradient echo is used. More information can be found in [79]. Reproduced from [81] with permission.

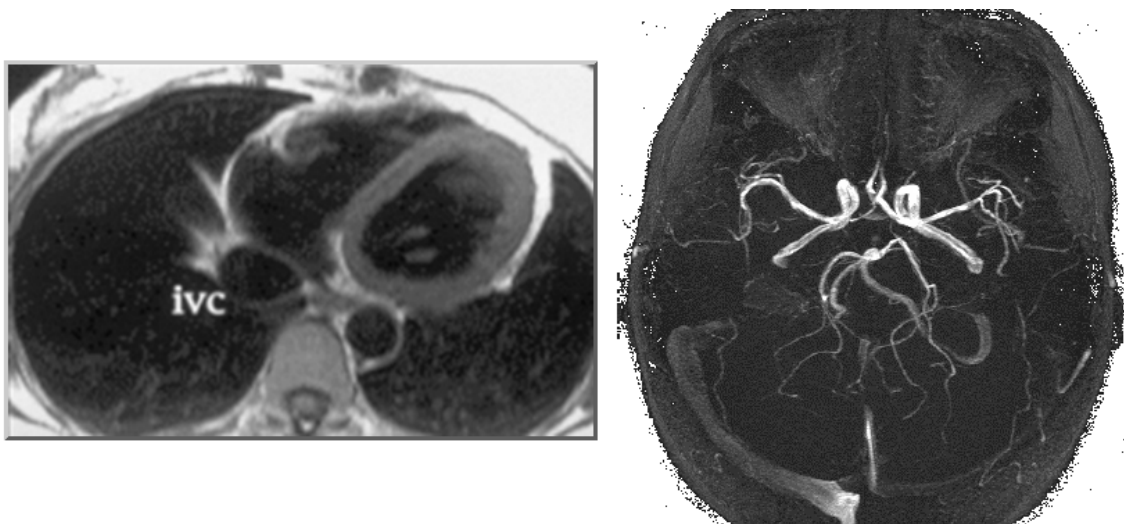


Figure 4.11: Black blood MRA (left) cardiac MRI of the inferior vena cava. Bright blood 3D MRA (right) of the intracerebral circulation. Reproduced from [79] with permission.

inversion recovery (DIR) [143], motion sensitive driven equilibrium (MSDE) [144] as well as the Delay Alternating with Nutation for Tailored Excitation (DANTE) [145]. Each has been successfully tested at 3 T or below but the lack of a body coil and an extended $\mathbf{B}_1$ coverage at 7 T requires the careful design of a dedicated RF coil.

DANTE

DANTE is a sequence preparation method for BB imaging where the module is added before excitation and readout [145]. It comprises non-selective low flip angle pulse trains in combination with interspersed gradient pulses with a short TR, figure 4.12. The result is preservation of the longitudinal magnetization of static tissue and attenuation of the signal coming from the flowing spins. In fact, blood spins cannot establish transverse steady state due to the spoiling effect induced by the gradient applied in the direction of the flow. A useful property of DANTE is that the signal suppression is largely insensitive to the velocity profile, which varies within the large cross section of the carotid arteries. Figure 4.13 shows the results of DANTE blood suppression at 3 Tesla.

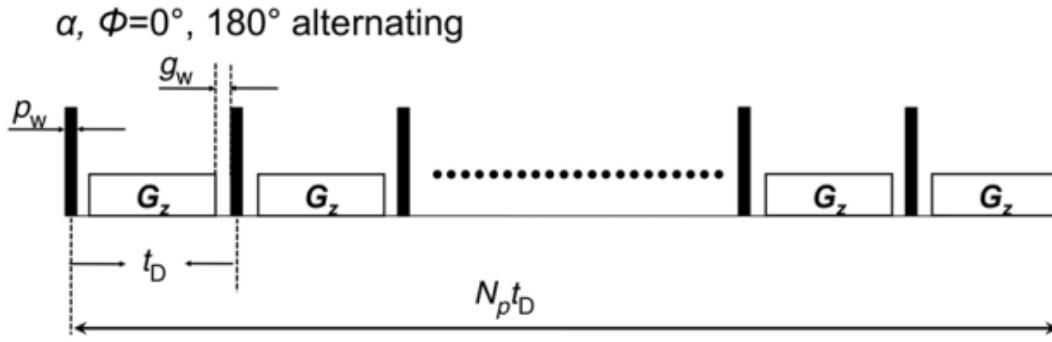


Figure 4.12: DANTE preparation module with non-selective RF pulses with flip angle from 3° to 15° , either with constant 0° phase or alternating phase between 0° and 180° . N_p is the number of RF pulses. t_D is the time interval between successive pulses, usually below 5 ms, during which a gradient is applied along the direction of flow. Typical values are: $p_w=50\text{-}200\ \mu\text{s}$, $N_p=150\text{-}200$ and $t_D=1\text{-}2\ \text{ms}$. Reproduced from [145] with permission.

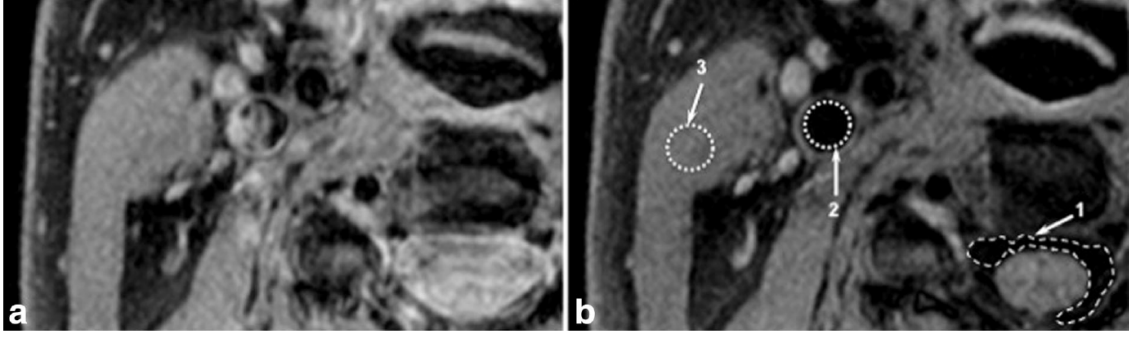


Figure 4.13: Efficiency of blood suppression is shown in these DANTE prepared 2D TSE images. In **a** the pulses were just 2, $N_p = 2$, and is clear that suppression is not achieved while in **b** $N_p = 1600$ with CSF and blood appearing black (circles 1 and 2) and static tissue retaining its signal intensity. Reproduced from [145] with permission.

This method will be used in this study in order to perform high resolution carotid artery wall imaging at ultra high field. The RF coil design, however, needs to be optimised based on the $\mathbf{B}_1$ field requirements of the sequence, due to the absence of body coils at 7 T.

4.3.2 Transmission Array Design

Typical values of the DANTE preparation pulses that have worked at 3 T as shown in figure 4.12 are:

- Number of pulses: $N_p=150$
- Pulse width: $P_w = 60\mu s$
- Inter pulse spacing: $t_D=1.5ms$
- Flip angle: $\alpha = 15^\circ$

The flowing spins need to be subjected to the above stated flip angle for the 150 pulses until they reach the bifurcation of the carotid artery. The blood velocity, v_{blood} , in the aorta is documented at 40 cm/s, [146], and an average velocity in the carotid arteries is a little bit less, at 35 cm/s. The required $\mathbf{B}_1$ field depth below the imaging plane can be estimated using equation (4.4).

$$L_{B_1} = v_{blood} \cdot N_p t_D = 40cm \cdot s^{-1} \cdot 150 \cdot 1.5ms \simeq 100mm \quad (4.4)$$

where L_{B_1} is the depth from the imaging plane (bifurcation) towards the aorta; v_{blood} is the blood velocity. For the sequence specifications prescribed in our application the value of L_{B_1} is thus about 100 mm.

The DANTE flip angle pulses can vary from $3^\circ - 15^\circ$ and for the most demanding scenario of 15° , the required $\mathbf{B}_1$ field strength can be estimated using equation (4.5). Based on this flip angle and the pulse width of $60\mu s$ the required $\mathbf{B}_1$ field strength is $16.3\mu T$.

$$\alpha = \gamma B_1 \tau \Rightarrow B_1 = \frac{\alpha}{\gamma \tau} = \frac{(15^\circ/360^\circ)}{42.6MHzT^{-1} \cdot 60\mu s} \simeq 16\mu T \quad (4.5)$$

where the flip angle is set at 15° and τ at $60 \mu s$.

The required DANTE pulse $\mathbf{B}_1$ profile could be regarded as a cylindrical shape with diameter of about 70 mm and 100 mm length. Within this cylinder an approximate $16\mu T$ B_1^+ is required, and it can be achieved with multiple surface coils. The $16\mu T$ requirement can be reduced to half by either doubling the duration of the pulses or reducing the flip angle and this is deemed more reasonable for reduction of RF heating. Thus, $8\mu T$ minimum $\mathbf{B}_1^+$ is regarded as a more reasonable target value.

For exciting the carotid arteries in the bifurcation two surface coils are placed at each side of the neck with another two surface coils placed on the lower part of the neck to extend the $\mathbf{B}_1$ to the required depth. For the proposed transmit array the following series of images from the simulations provide the reasoning behind the selection. Figures 4.14 and 4.15 provide a visualisation of the setup based on anatomically realistic models from the virtual population [147], on an electromagnetic simulation platform, Sencad-X, (Speag, Zurich), with the proposed coils in position.

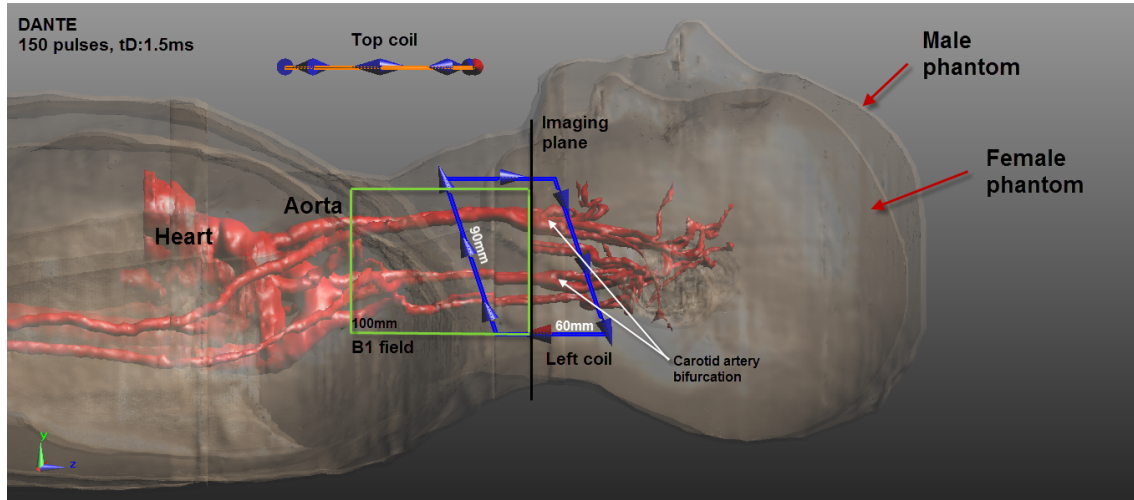


Figure 4.14: Duke and Ella from the virtual family are overlaid with the cardiovascular system highlighted to indicate the procedure of the coil positioning. Left and Right coils are used for suppression and imaging excitation but the top coil is used only for the DANTE pulses to extend the profile to the required depth. The side coil dimensions are 90 mm $\times$ 60 mm. For 150 pulses and 1.5 ms inter pulse spacing the B_1 field requirement is marked with the green square.

The particular coil shape used was as such to cover the RoI as much as possible. Simpler shapes like plain or square loops were considered with either copper wire or tape. They were evaluated and did provide a good field cover but the shape lacked in optimisation of coverage and mechanical fit with the main cover. Plain loops however were used for the receive array as it will be explained in the next section. Since multiple smaller loops were necessary in the Rx array the plastic cover to hold the assembly was designed appropriately.

Figures 4.16, 4.17, 4.18 and 4.19 provide a representation of the B_1 field for channels 1, 2, 3 and 4 respectively. The field is shown only for the region of interest, which is the carotid arteries. Also, the fields are normalised based on the maximum power provided per channel by the amplifier of the system (1000W per channel). Finally, the field is shown in the figures for values higher than $8\mu T$ which would generate the required excitation. Anything less than $8\mu T$ is coloured in black.

Figures 4.20 and 4.21 show the total combined B_1 field for a selected RF shim. The first image maintains the colour variation to depict the magnitude and the

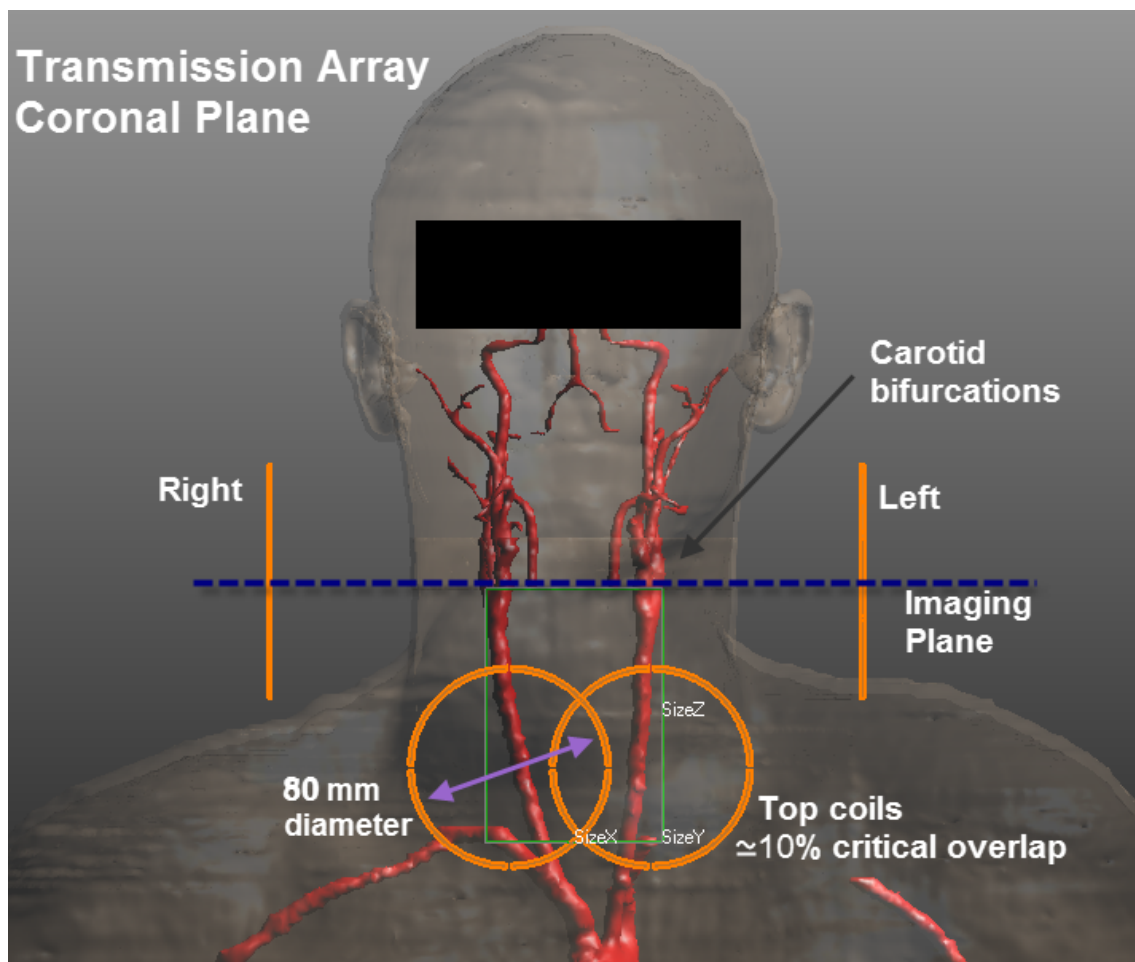


Figure 4.15: CAD top view of the proposed transmit array. The two top circular coils are positioned lower than the imaging plane (blue dotted line) for the extension of the $\mathbf{B}_1$ profile required for DANTE BB preparation. The overlap of the two coils is used for minimisation of the interactions between the coils and to optimise their stability. Their diameter is selected to be 80 mm for adequate coverage.

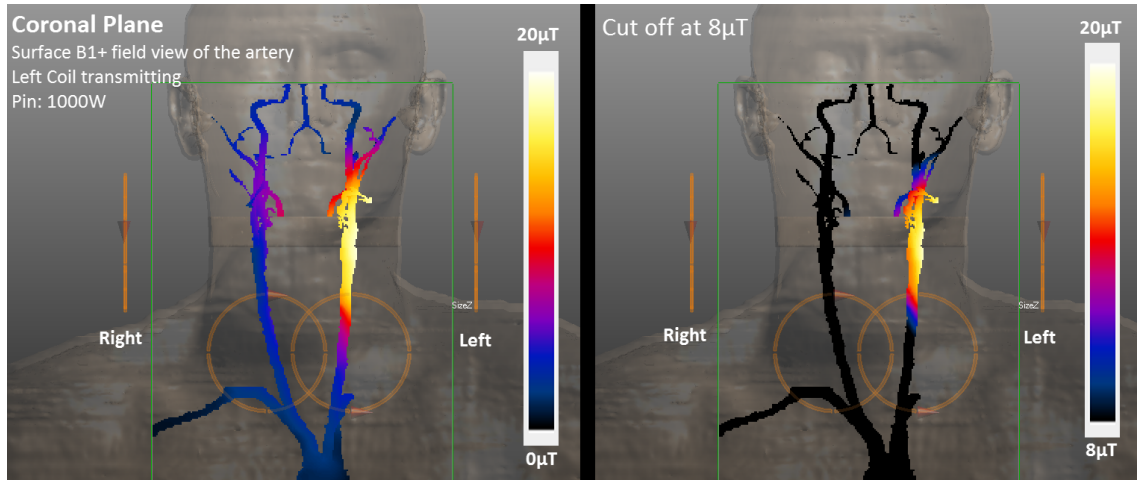


Figure 4.16: Coronal plane of the B_1^+ field of the left coil transmitting at the maximum power provided by the pTx system RFPA (1000 W). The arterial part of the model has been isolated in the viewer. The image on the left is the field scaled from 0 to $20 \mu T$ while the right image is based on the calculated minimum $8 \mu T$ cut off. Cut off image shows that there is coverage from the imaging plane level until about the uppermost of the top coils.

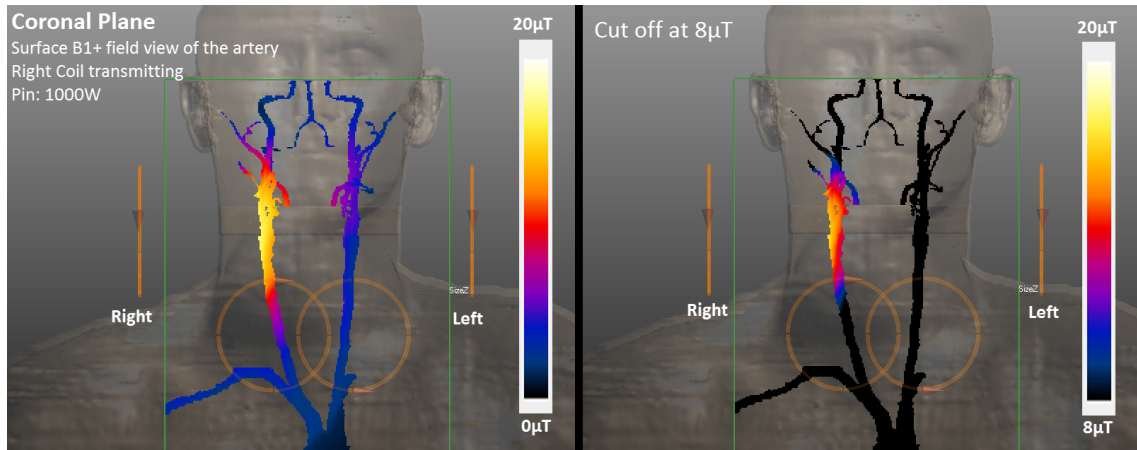


Figure 4.17: Parameters as for previous image with the right coil transmitting at full power and cut off at $8 \mu T$ shown on the right. Sufficient coverage in the imaging plane is achieved and is symmetrical with the left coil coverage.

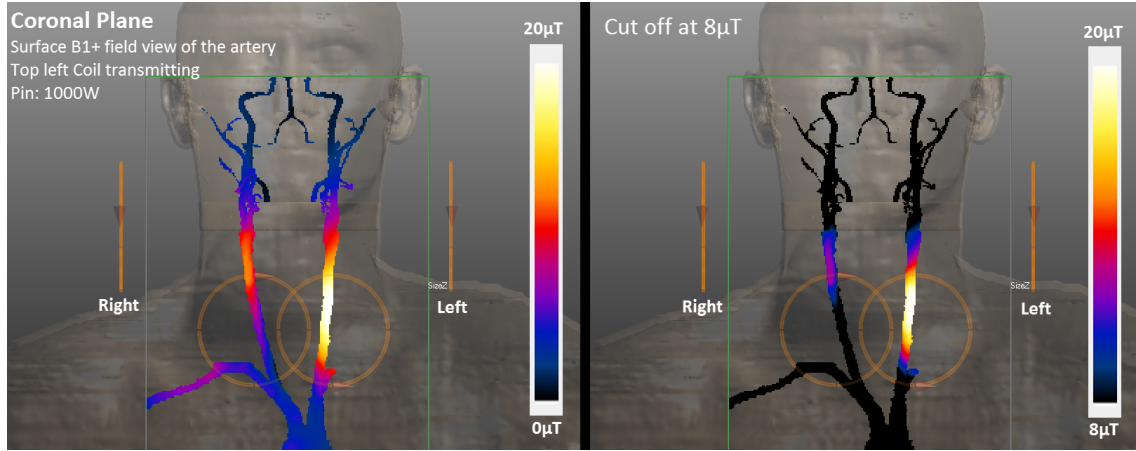


Figure 4.18: Top left coil coverage illustration with the same parameters as shown in the previous images. These coils will be useful only for the DANTE train preparation part of the sequence and hence the $8\mu T$ is more important to be met. Coverage of this coil seems to complement conveniently with the side coils extending the field continuously.

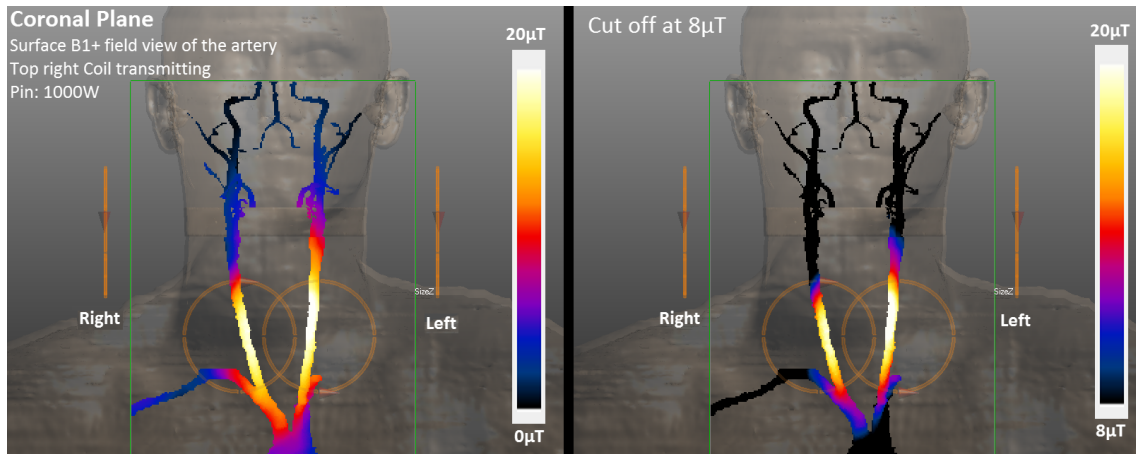


Figure 4.19: Top right coil showing clear coverage in the cut off image of the lower part of the carotid artery.

second image is an iso-surface of the RoI for $8\mu T$ and above. For both plots it is clear that coverage is adequate and starts from well within the 100 mm length and maintains strong and adequate $\mathbf{B}_1$ all the way to the bifurcation. Based on the findings in these figures, a coil constructed with these specifications should generate an RF field strong enough in magnitude and coverage to match the required DANTE prepared sequence and achieve sufficient blood suppression. The construction of the transmit array as designed here will be realised in the next section.

To sum up, four Tx array elements are designed with the side elements having a long dimension ($90mm \times 60mm$) covering effectively the two sides of the neck. Also, the $\mathbf{H}$ -field is sufficient to include an average anatomical variation of the carotid arteries. The top coils were selected at a diameter of 80 mm to adequately extend the field from the side coils down to 100 mm below the imaging plane. The transmit array will be realised in the next section. The following stage, however, is the consideration of the more sensitive part of the RF coil which is the receive array.

4.3.3 Receive Array Design

Signal-to-Noise Ratio

Historically, volume solenoids or homogeneous resonators were used for signal reception, but it has been shown that if the location of interest is known, surface coils can be used for signal-to-noise ratio (SNR) improvement [67]. A local coil acts as a spatial filter removing the broadband thermal noise generated by the subject in the areas away from the RoI. The calculation of the SNR in an NMR experiment is well documented and can be found in [148, 149, 150, 151]. In particular, the electromotive force (emf) induced by the presence of a rotating magnetisation, M , in the sample is related to the sensitivity profile of the coil, $\mathbf{B}_1^-$, see equation (4.6).

$$emf = - \int_{sample} \frac{\partial}{\partial t} \mathbf{B}_1^- \cdot M dV \quad (4.6)$$

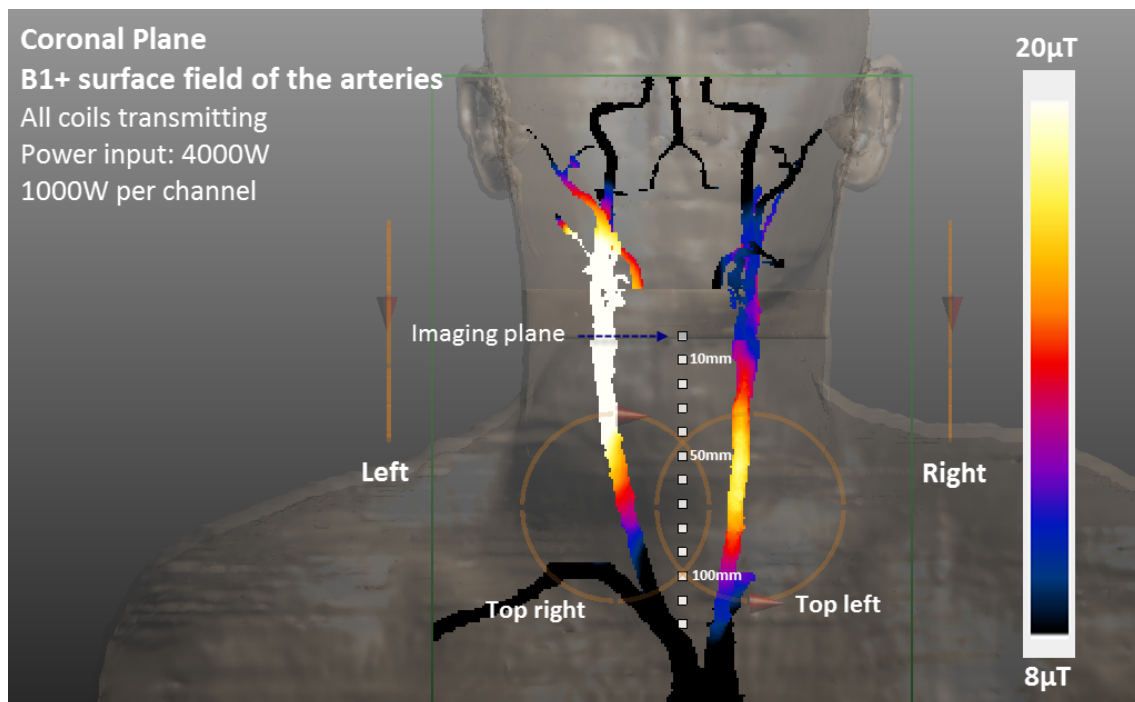


Figure 4.20: In this image all coils transmit with the available power per channel with all channels transmitting with same magnitude and phase. The image scale is from $8\mu T$ to $20\mu T$, i.e. all values below $8\mu T$ are black. The depth points also show how the coverage relates to the required 100 mm from the imaging plane with sufficient observed field coverage.

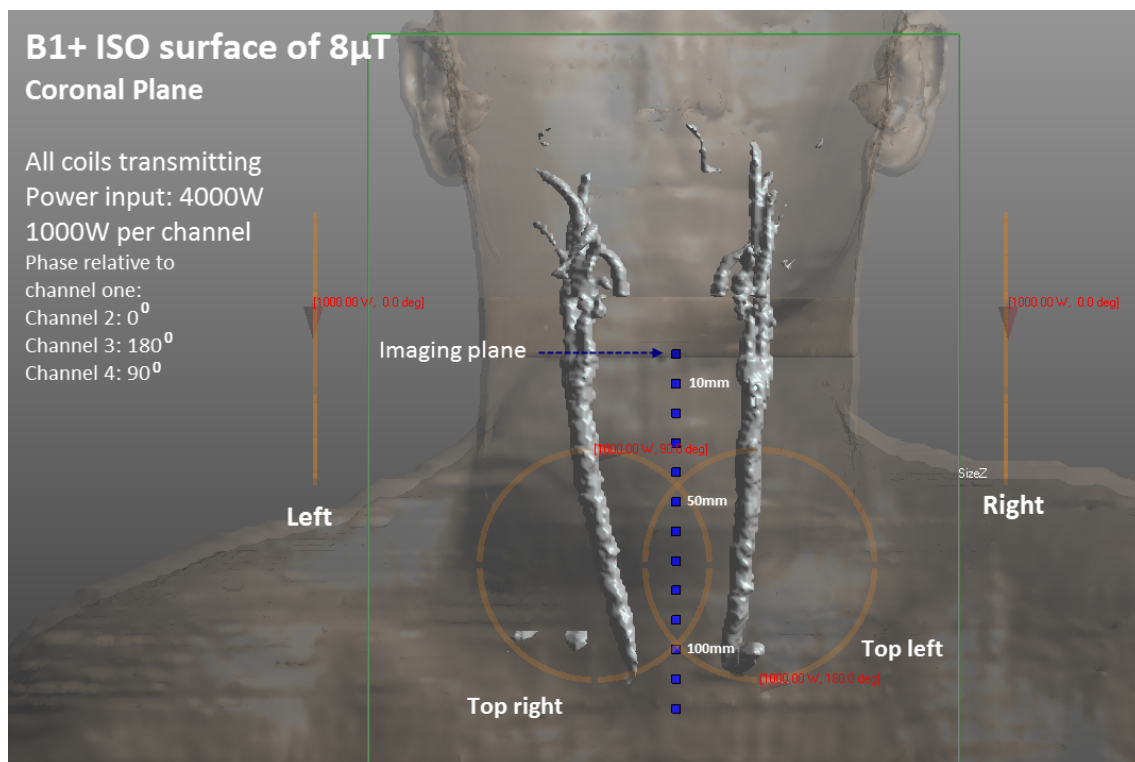


Figure 4.21: An appropriate RF shim as shown on the image is selected and the surface that exceeds $8\mu T$ is highlighted in this ISO surface image. It can be concluded that it is within the array's capability to cover the required area with the desirable RF field strength.

where $\mathbf{B}_1^-$ is the coil magnetic flux density per unit current and M is the magnetisation of the spins. For homogeneous samples the spatial dependence of M can be factored out and the general conclusion is that the received signal is proportional to the sensitivity $\mathbf{B}_1^-$.

$$S \propto \mathbf{B}_1^- \quad (4.7)$$

Now, the electric noise generated in the probe is:

$$N_p = \sqrt{4kT\Delta f R_{in}} \quad (4.8)$$

where k is the Boltzmann constant, T is the sample temperature in Kelvin, Δf is the receive bandwidth and R_{in} is the equivalent coil resistance.

Thus the SNR is in fact proportional to:

$$SNR = \frac{V_{signal}}{V_{noise}} \propto \frac{\mathbf{B}_1^-}{\sqrt{4kT\Delta f R_{in}}} \quad (4.9)$$

Equation (4.9) shows that the SNR is reliant on the optimisation of the probe sensitivity profile as well as the minimisation of the equivalent resistance of the coil. For the latter a reference has already been made in section 4.2.2 but in short it can be broken down to:

$$r = r_\Omega + r_M + r_E \quad (4.10)$$

where r_Ω represents the ‘ohmic’ resistance introduced by the unloaded coil (the skin effect needs to be taken in to account for its calculation); r_M represents the magnetic losses due to the conductivity of the sample, i.e. the time varying magnetic flux creates eddy currents in the sample; and r_E represents the dielectric losses which are an intrinsic property of any material when subjected to an $\mathbf{E}$ -field. The radiation losses can also be included in the last term.

A thorough discussion has been made in section 4.2.2 on how to address some of these noise sources. In particular radiation and dielectric losses are minimised

by balancing, the use of shield traps and capacitance segmentation and they will be addressed again in the physical construction of the probe. The magnetic losses are inevitable due to the necessary $\mathbf{H}$ -field and can not be reduced unless the sample properties are altered. However, by placing the receive coil closer to the sample the sensitivity profile is maximised which increases the SNR.

Optimal Array

A coil can operate in transmit/receive mode where a TR switch will interchange between the RF power amplifier and the RF receiver. However, the positioning and geometry of the coil has to be designed in order to approximate the best performance for each function. Even though the TR solution is the simplest, the most efficient method is to use a separate receive array that consists of multiple elements that surround the sample [152, 153, 154]. Once the preference for surface coils is established, the next stage is the definition of the size, shape and material to be used. There is not a definite answer for these parameters and in fact they can vary for similar target applications.

The selection of the element dimensions becomes more complex to approach mathematically when a multi-channel array is used. However, an array is preferred for its documented efficiency since it combines the SNR improvement of local surface coils for a larger area. The receive arrays can be used for SNR increase by optimised combination of the signals from each channel or for the reduction of scan time by utilising the spatial information of the distinct sensitivity profiles, a.k.a. parallel imaging. How this is done is not referenced here but more information can be found in [155, 156, 157, 158]. The simplest way to reconstruct images using a receive array is to use the Sum of Squares (SoS) method where for each voxel the signal from each loop (weighted if necessary) is squared and averaged with the final image being the square root. The result image is an averaged overlaying of all the images individually.

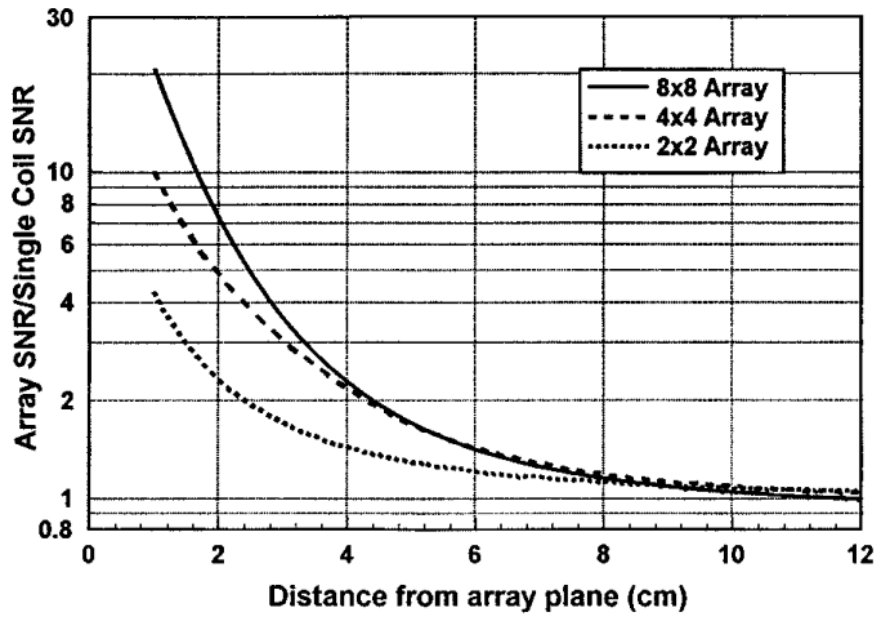


Figure 4.22: Signal-to-noise ratio (SNR) of various loop arrays versus a single coil with the same dimension. SNR was calculated on the axis of the array extended in the sample. The benefit from using multiple loop arrays is striking when the array is positioned close to the sample. Reproduced from [152] with permission.

Figure 4.22 shows that it is possible to improve the SNR by using multiple receive elements at the same time. The diagram shows a comparison of a square loop with the same overall dimensions as the arrays with either 4, 8 or 16 elements. When positioned close to the sample, the SNR improvement is apparent; it also shows that the SNR never falls lower than the SNR achieved by a single loop with the same dimensions. In figure 4.23 the particulars of the investigated application are illustrated. A male phantom with the arterial system highlighted is shown in the simulation environment and the coil positioning drawn at each side. The distance between the proposed array positioning and the inner side of the carotid bifurcations, plus a margin for anatomical variation, is less than 50 mm, where the loop diameter of the element is selected at 50 mm. Thus, an SNR improvement is expected based on the number of elements in the array.

Due to the anatomical details of the region of interest, instead of opting for a semi-spherical array shape as would be done for the head, the four elements are

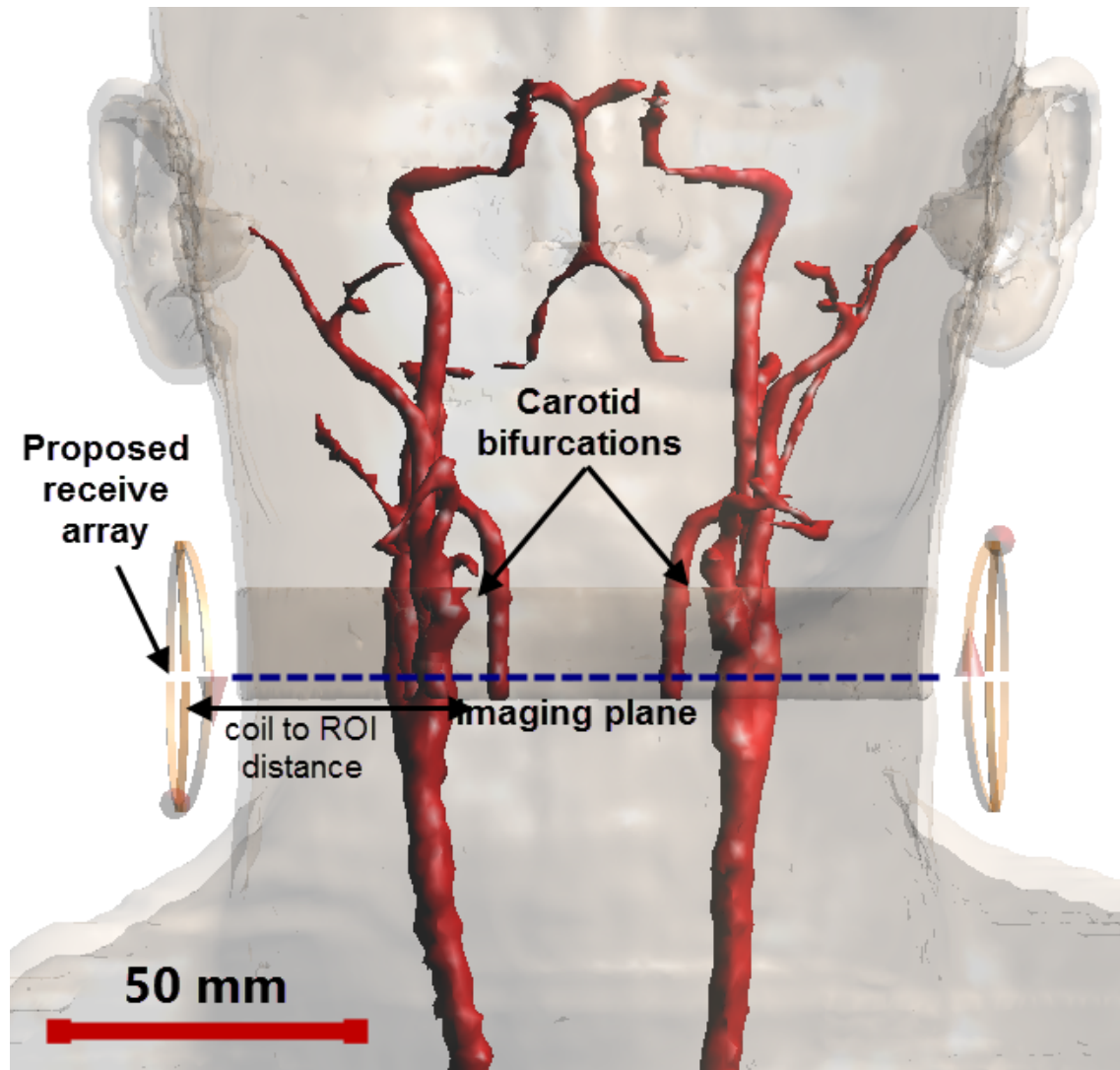


Figure 4.23: The imaging plane is marked with the blue dotted line and is the axis for the positioning of the proposed array. The distance from the centre of the array to the carotid arteries plus some variation margin to account for different subjects is approximately 50mm. Based on the diagram in figure 4.22, the SNR improvement vs loops of the same size would be roughly 50%.



Figure 4.24: Illustration of the proposed receive array. Two loops of 50 mm diameter are positioned at each side, overlapping for decoupling.

equally positioned to cover the two sides of the neck based on the fact that it is the closest point from the skin to the arteries. The number of elements selected for this application was limited by the anatomical difficulty of the region where space is limited for placing many large loops between the shoulders and the head. Figure 4.24 illustrates the 4 channel proposed array and figure 4.25 shows one side of the array.

The optimal SNR when SoS are used can be seen in figure 4.26. In short, a separate receive array is proposed here which can be placed very close to the skin and benefit from the multiple elements for SNR improvement. The dimensions have also been optimised for the particular region of interest. In the next section, the construction of both the transmit and receive arrays will be realised.

4.4 RF Coil Construction

4.4.1 Computer Aided Design

Figure 4.27 shows the proposed coil in a CAD software environment. The casing of the coil was designed so that there is enough space for the accurate positioning

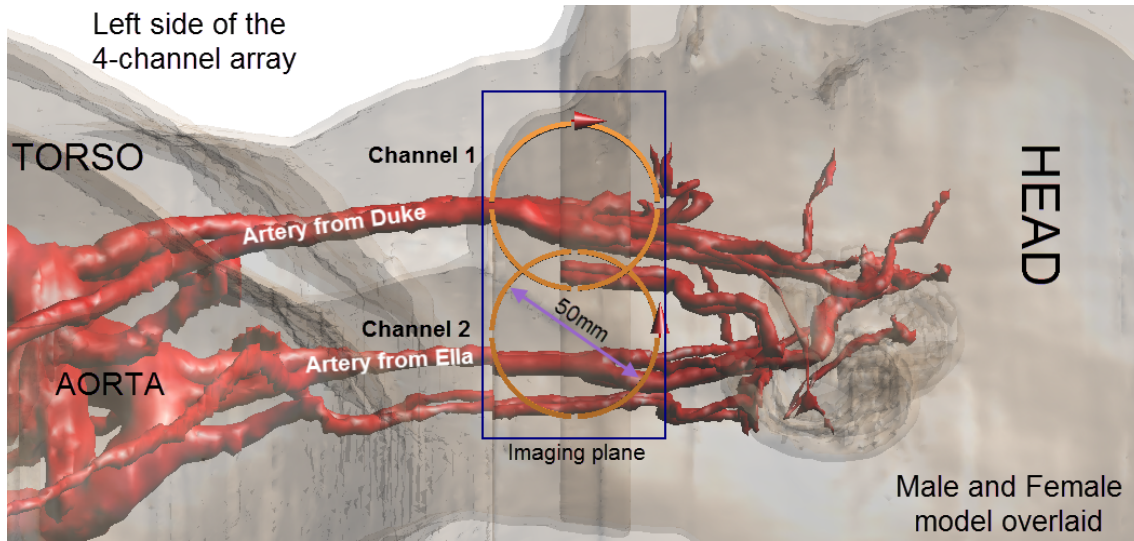


Figure 4.25: One side of the proposed array. The illustration includes Duke and Ella phantoms overlapped with the arterial system highlighted. The coils are positioned on the imaging plane, i.e. at the bifurcation level. The size and positioning will produce a side sensitivity that will cover an average anatomical variation from subject to subject. The diameter was selected to be 50 mm for a slight SNR benefit at an average carotid depth.

of the loops and the associated electronic components as well as to comfortably fit an average human subject. In figure 4.28 the coil is shown with and without the plastic cradle loaded with a human phantom. The top compartment hosting the two top transmit loops is removable so that the subject can lie between the left and right side of the coil. This compartment is connected via coaxial cabling and BNC connectors to the main cradle for feeding the RF and the DC logic signal for the PIN diodes. In figure 4.29 the final coil version can be seen.

4.4.2 Decoupling Networks

It is clear by now that the coil has been designed to operate interchangeably; in transmission mode with the transmit array; in receive mode with the receive array. Both arrays need to be electromagnetically separated when in their respective mode keeping the relevant one tuned at the Larmor frequency while the other is detuned. Apart from the necessary decoupling between the transmit and receive arrays, the loops within the arrays have to be decoupled relative to each other. A coupled loop

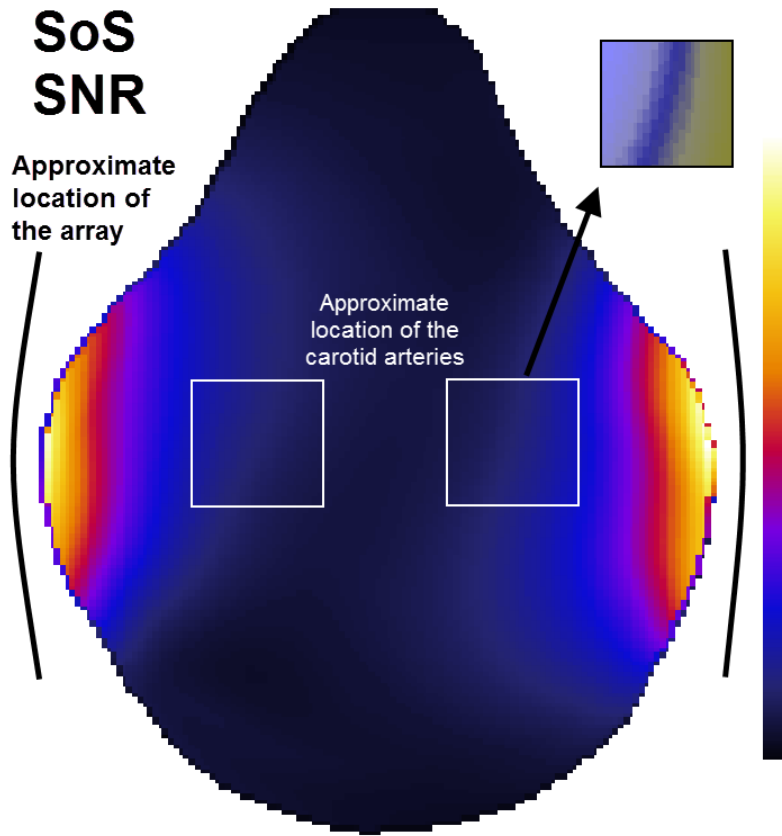


Figure 4.26: Use of Sum of Squares (SoS) method as defined above for the estimation of the SNR profile. The scale is relative showing the sensitivity profile and how it decays rapidly for areas deeper in the neck.

impedance is increased by $M_k I_k$ with M_k being the coupling coefficient and I_k the current flowing in the other loop(s), see section 4.4.2. The latter increases the R_{in} of the coil that defines the total noise, lowering the SNR of the receive array.

Parallel imaging methods rely on the distinction of the sensitivity profiles of each of the elements. The SNR is intrinsically reduced by the use of such methods due to the geometrical arrangement of the array. However, the SNR can be further reduced if the sensitivities from each loop are correlated thus introducing ambiguities. Coil decoupling acts to optimise the performance of any parallel imaging method and hence it is desirable.

For the above mentioned reasons the coil has the following decoupling systems embedded:

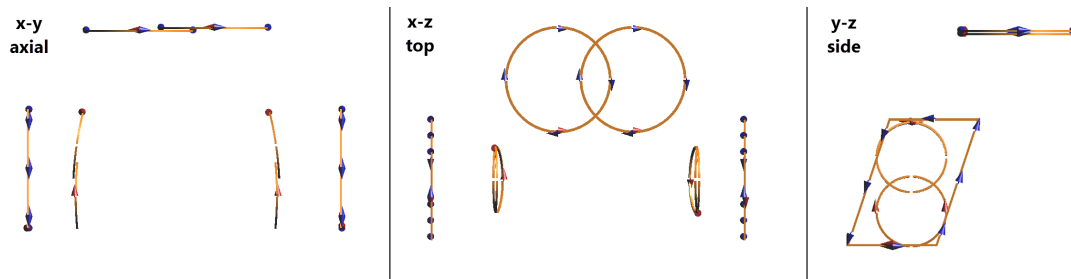


Figure 4.27: Full array illustration in the three planes without phantom or any cradle. Based on this model an appropriate plastic cover was determined, designed using CAD software and 3D printed.

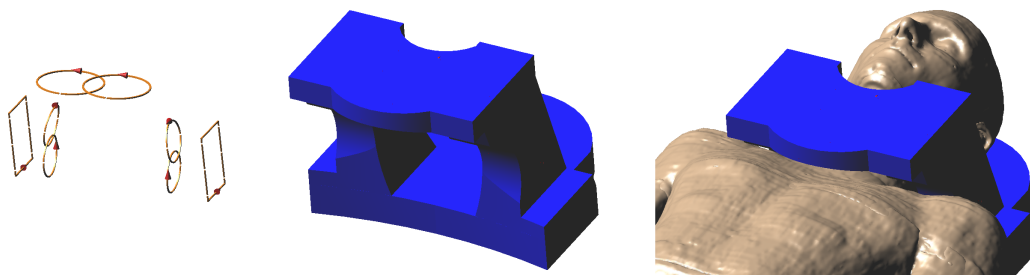


Figure 4.28: The plastic cover as shown in the middle image is anatomically designed to fit an average neck. The top compartment hosting the two top coils is removable for the positioning of the subject. Some parts were constructed by the use of a high precision 3D printer and for the base, plastic sheets were CNC milled.

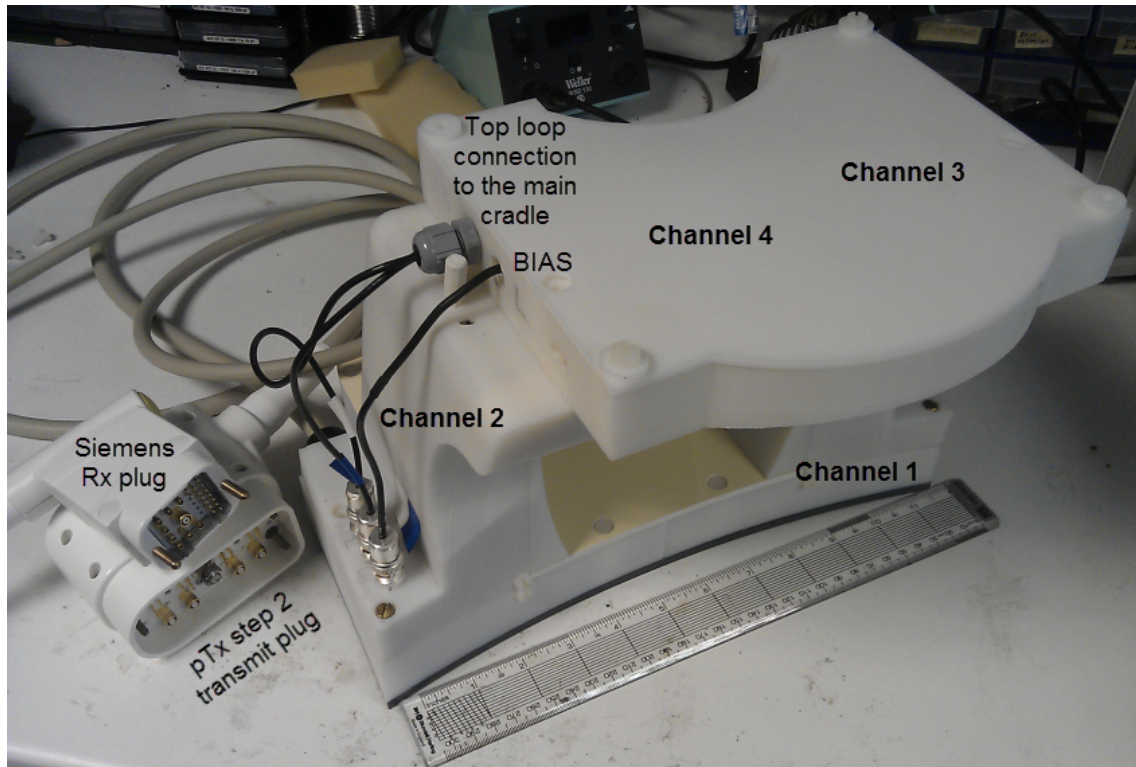


Figure 4.29: Plastic case of the coil 3D printed and fitted with the receive and transmit arrays. The two plugs can be seen on the side, the trio plug for the Rx elements and the pTx-step 2 for the Tx array. The top coil is detachable so that the subject can be positioned. The top is connected to the main cradle with BNC cables for the RF transmission and the biasing of the PIN diodes.

- Geometric decoupling by cancelling the magnetic flux by overlapping the loops.
- Active detuning network for the transmission and receive array controlled by PIN diodes.
- Passive detuning network for the receive loops.
- Pre-amplifier decoupling, a method that further isolates the coupling between receive loops in an array.

Geometric Decoupling

Geometrical decoupling was employed for both the receive and transmit arrays in order to reduce the coupling, [154]. The two loops are tuned at the Larmor frequency and then brought towards to each other until the characteristic resonance splitting

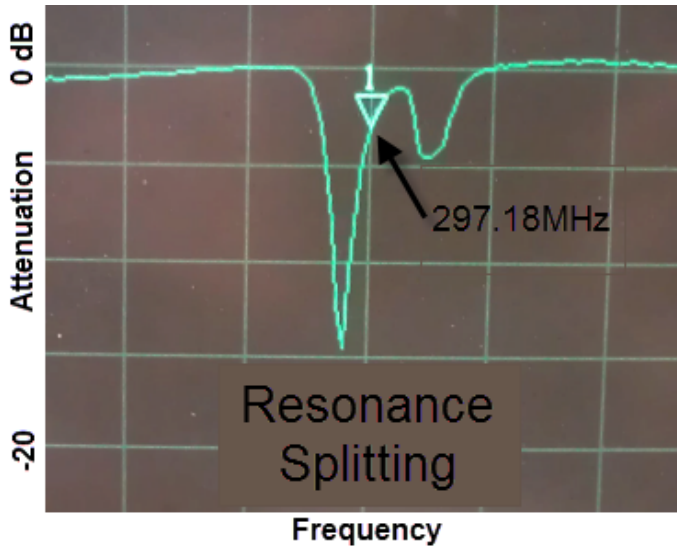


Figure 4.30: Resonance splitting is an effect observed when there is coupling between two individually tuned loops. If no decoupling measures are taken it is not possible to re-tune both coils.

completely disappears, see figure 4.30. Figure 4.31 shows the overlap of the two loops for the left side of the receive array with the corresponding S_{12} measurement from the network analyser. Figure 4.32 shows how the top two loops of the transmit array have been positioned.

Active Detuning

Active detuning was performed with the use of high power, fast switching non magnetic PIN diodes. The components used for the coil construction are the Temex DH 80106 which can withstand 500 W power. The mechanism and typical performance of the detuning can be seen in the reflection measurement shown in figure 4.33.

The active detuning of the transmit array was performed with a pole insertion, i.e. an LC circuit added in series to the circuit. This method can be used for a multiple tuned coil but here it is utilised for effective detuning. As can be seen in the transmission array schematic in figure 4.34 the diode is connected in line with an inductance which, along with the fixed capacitor, creates an LC resonant circuit at the Larmor frequency. A photo of the left transmit loop can be seen in figure 4.35.

The electromagnetic decoupling of the receive array during Tx mode was done by applying DC biasing pulses to the diodes and short circuiting the tuning capacitor.

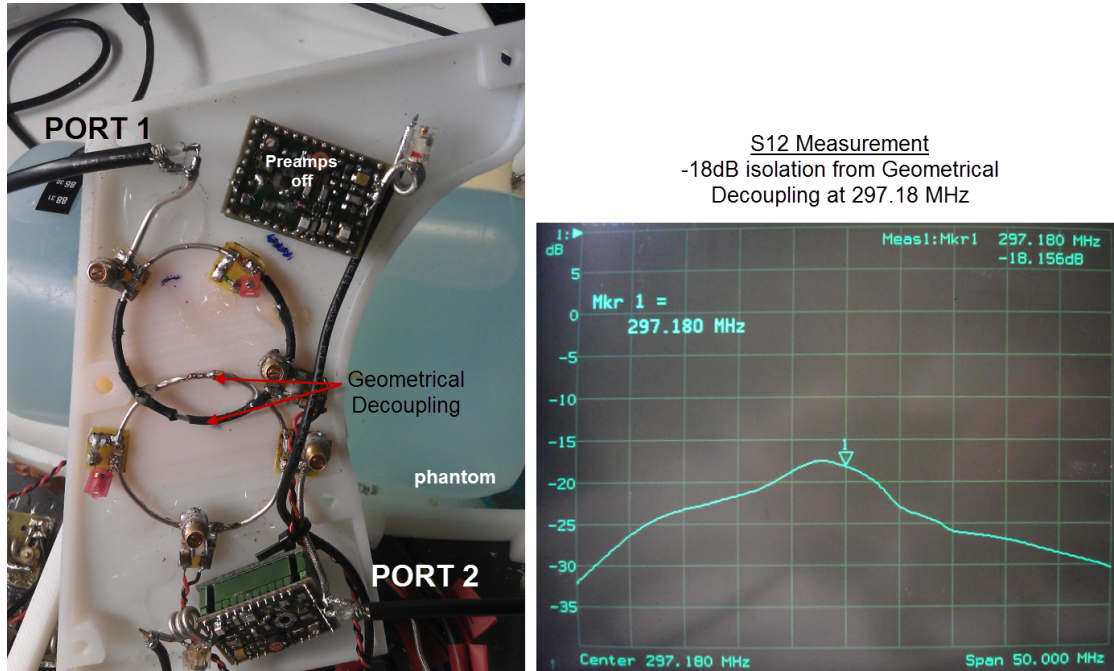


Figure 4.31: Geometric decoupling of the one side of the receive array. The coils are positioned carefully when an S_{12} measurement is taken. The -18.2 dB value is deemed adequate and the position of the loops is fixed with glue on the plastic cover.

A circuit diagram is shown in figure 4.36 and the corresponding photo of the left side of the receive array is shown in figure 4.37 with the active detuning stages marked with blue squares. In the same image passive decoupling is highlighted. This will be discussed in the next section.

Passive Decoupling

Passive decoupling is a circuit added on each of the loops in the receive array for more isolation during transmission mode, [159]. The right and left Tx channels are placed directly behind the receive array and during transmission; even after active detuning of the Rx, the remnant weak coupling can lead to some induced currents in the small loops if the RF pulse is powerful enough. Also, there is a possibility that the active detuning malfunctions and that the coupling becomes strong with significant currents that can alter the estimated RF heating.

A crossed pair of diodes is therefore positioned along with a small inductance as shown in the circuit diagram in figure 4.36 and in the photo in figure 4.37.

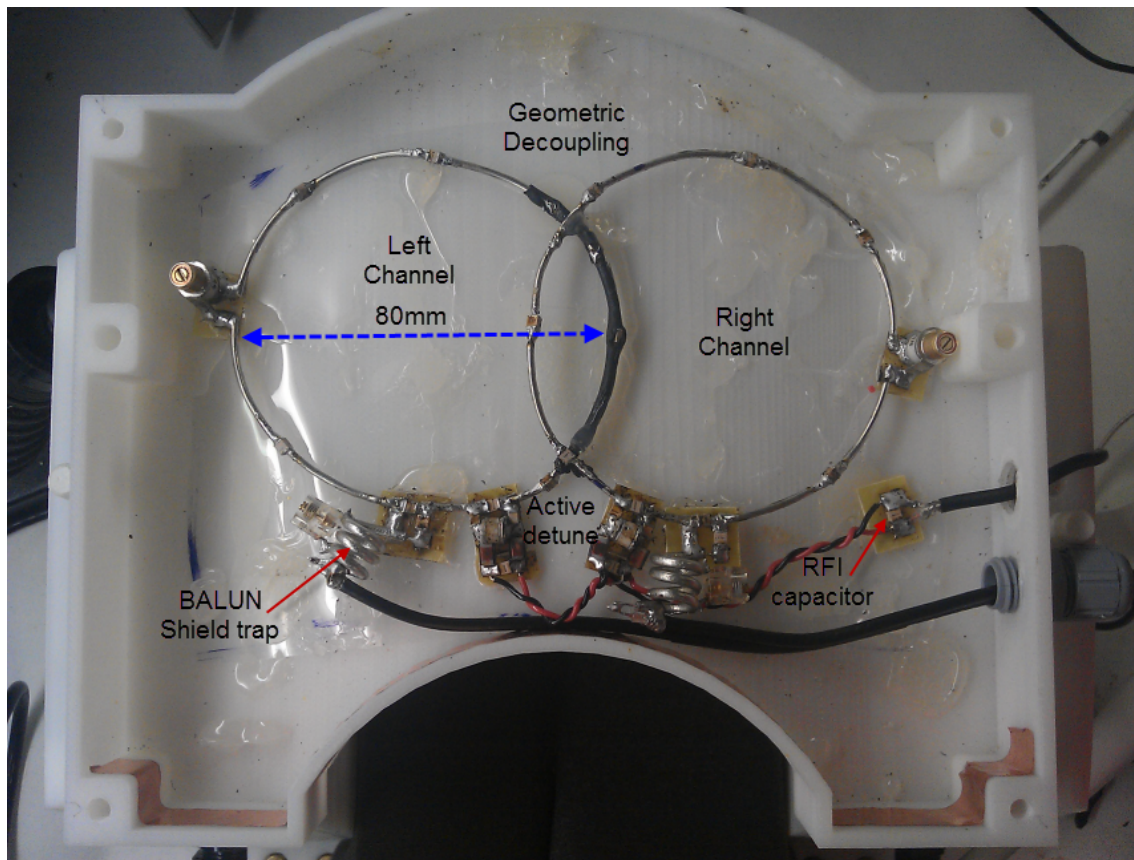


Figure 4.32: The same decoupling scheme was used for the two top coils. The decoupling in this case was -9.9 dB. All the coupling between elements for all coil states is summarised in Chapter 6.

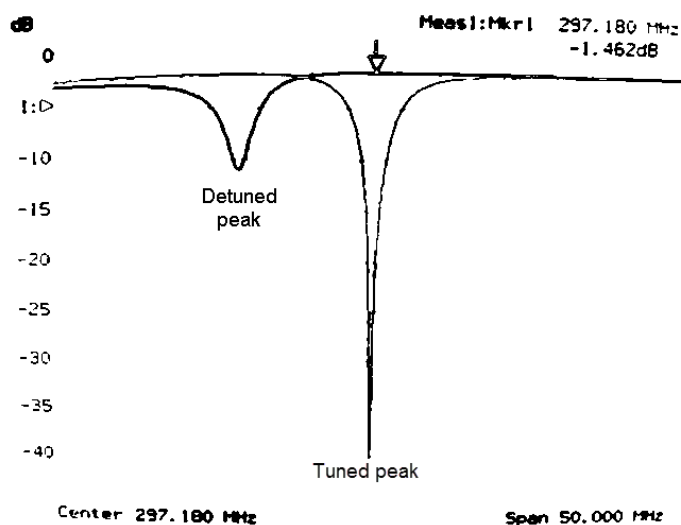


Figure 4.33: Network analyser S_{11} measurement of the two states for the side Tx element. The tuned high Q peak is switched with the pole insertion to a lower resonance leaving the coil unreceptive at the Larmor frequency. The full decoupling performance of the detuning stages is evaluated in Chapter 6.

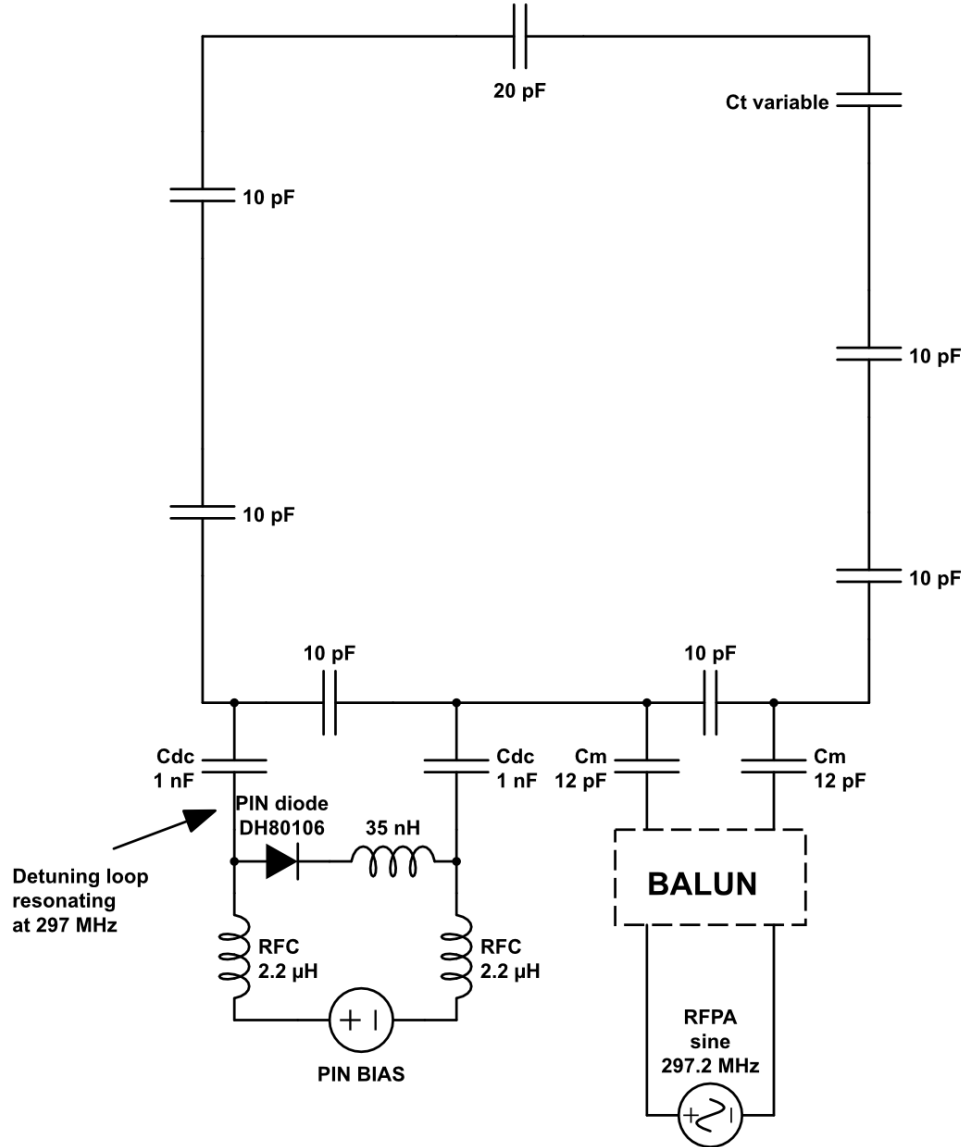


Figure 4.34: Circuit diagram schematic of the transmit loop. Each loop is designed to handle high power RF pulses and hence a balanced matching network is selected, i.e. the termination that would end on the grounded shield is connected with a matching capacitance of double the value in order to electrically raise the normally ground terminal and reduce the electric field potential. The PIN diode detuning network is also tuned at the Larmor frequency so that currents are suppressed by the high impedance introduced in series during receive mode.

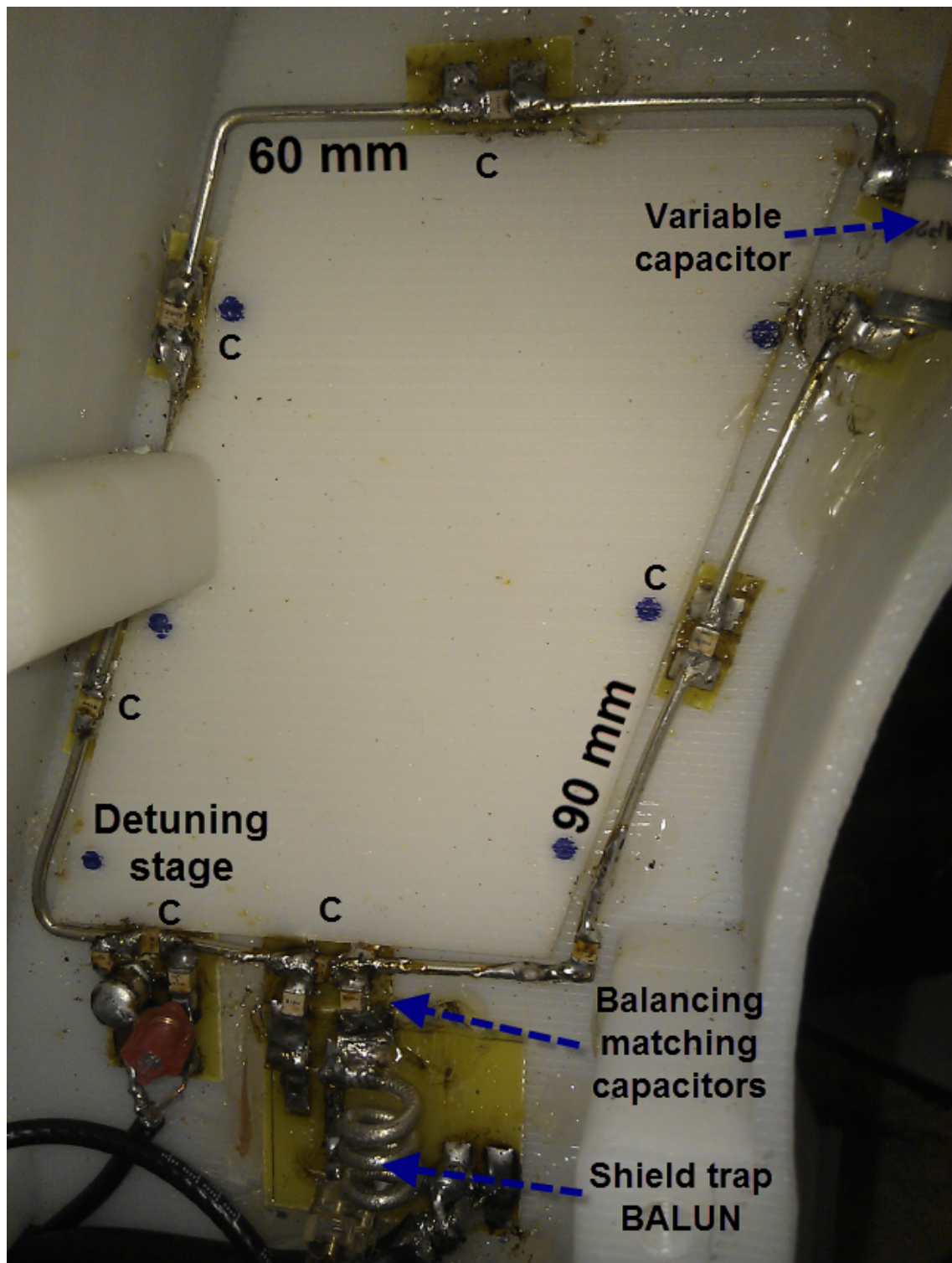


Figure 4.35: Photo of the finished left transmission loop. The detuning stage comprises the PIN diode, RF chokes, DC isolating capacitors and the inductance (orange covered component). The feeding and matching network comprises the balancing capacitors of the same value and the shield trap for reduction of the coupling between coil and shield. The variable capacitor is positioned on the top right and is controlled externally by a plastic rod.

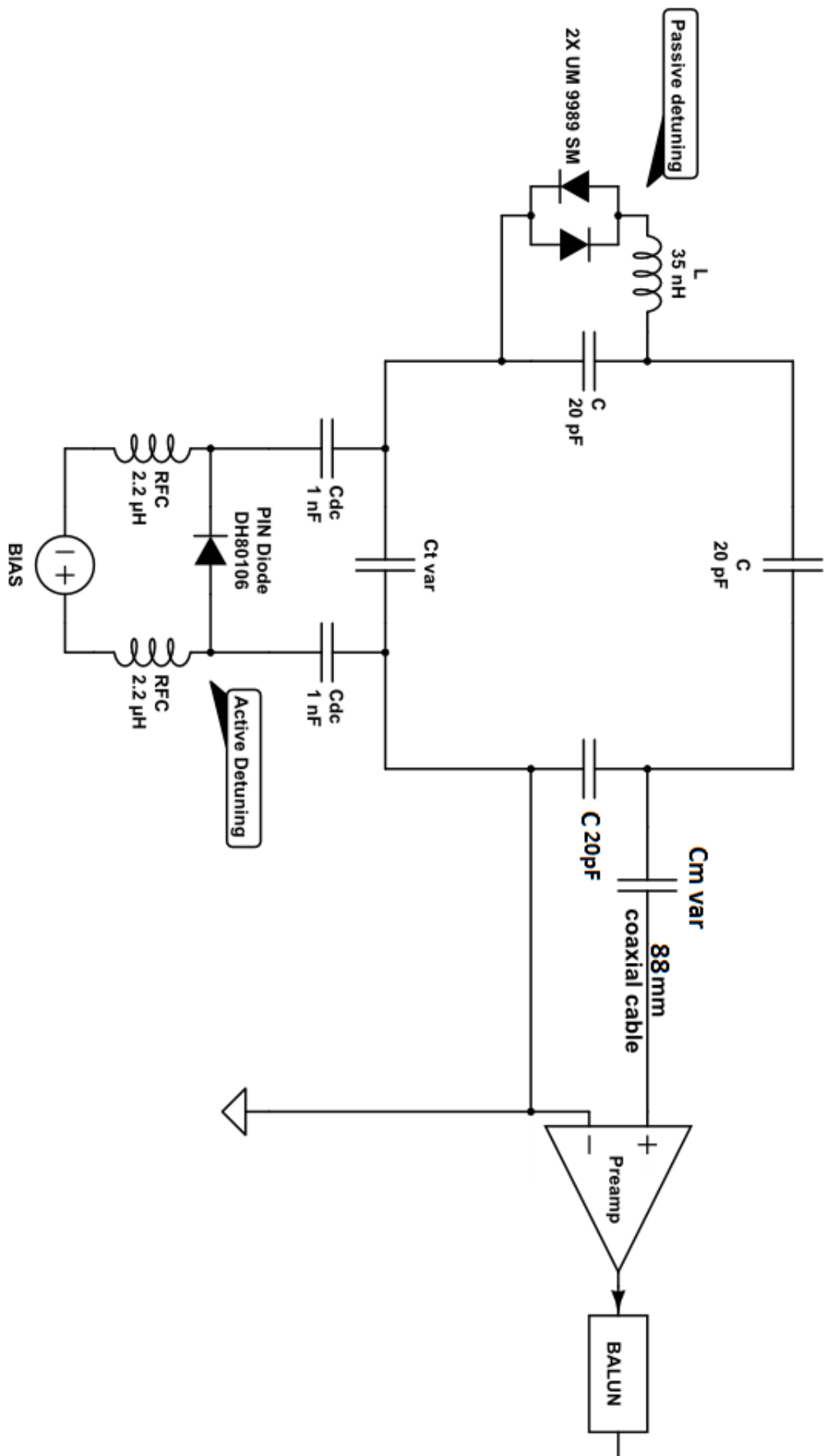


Figure 4.36: Circuit diagram schematic of the receive loop. It is imperative that there is adequate isolation between the transmit and receive arrays during transmission mode in order to minimise flowing currents in loops positioned by the neck skin. There are two stages of de-coupling that act in combination, reducing the chance of a failure. The active detuning, PIN diode DH80106, removes the Ct capacitor and detunes the circuit. At the same time, when the crossed diodes, UM9989, are biased by an RF pulse the resonant circuit is added in series blocking any currents at ω_0 .

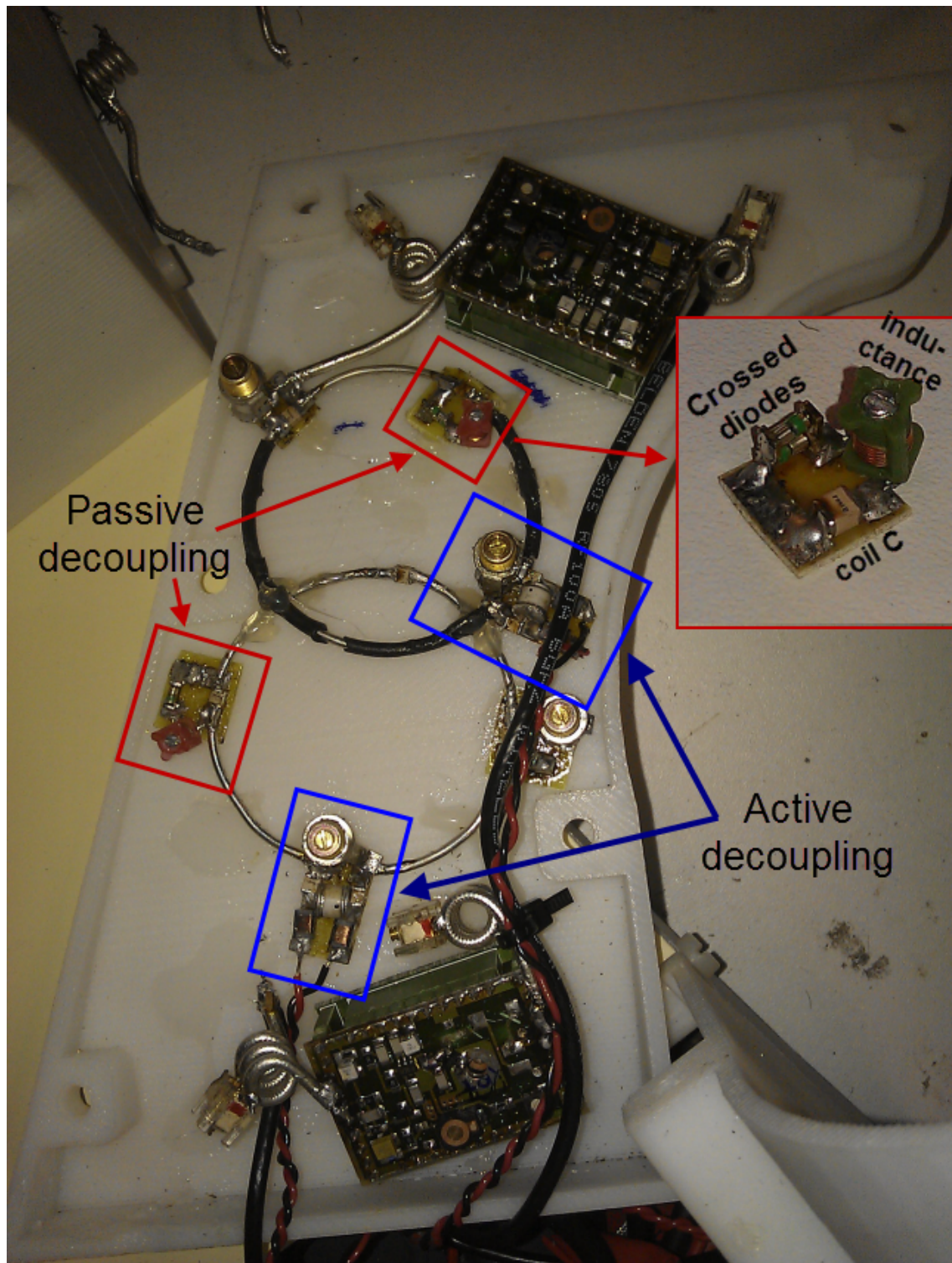


Figure 4.37: Photo of the finished left side array. The active detuning is visible here in blue. It is the same as in the transmission without the inductance for pole insertion. In red squares the passive decoupling is shown also in a close-up. The two sensitive crossed diodes are biased when the transmit coil is in operation. The total decoupling between the two arrays is evaluated in Chapter 6.

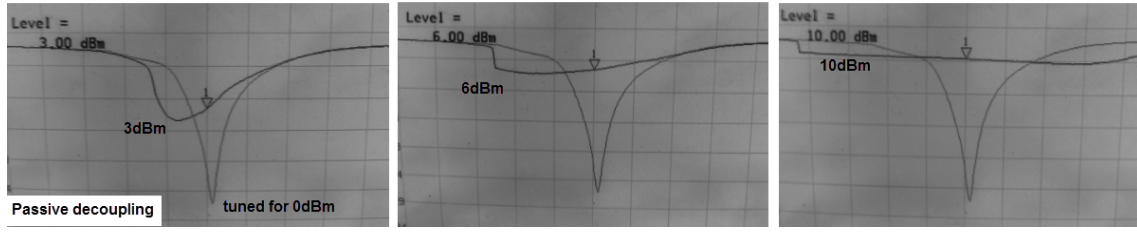


Figure 4.38: Passive decoupling testing. In the far left image the initial tuned peak is shown along with the new peak when a significant power is used for the S_{11} measurement. The 3 dBm partially biases the crossed diodes creating a distorted profile. The effect is further visible at 6 dBm and 10 dBm where the stage is activated, detuning the loop completely. Passive and active decoupling provide a sufficient electromagnetic isolation between the two arrays.

The particular diodes used here, Microsemi UM9989, have a very low forward bias threshold so that they are in the 'on' state for any reasonably high power RF pulse, inserting in this way an inductance which, in combination with the capacitance of the coil, forms a current trap at the Larmor frequency. During receive mode the induced voltage in the loops is very small and insufficient to bias the diodes.

A test could be performed with the network analyser showing that for an increase in the power input for the S_{11} measure the diodes are activated and detune the loop. Figure 4.38 shows how a tuned loop becomes detuned by the increase of the network analyser power from 0 dBm to 10 dBm.

Pre-amp Decoupling

The output voltage of any coil when in the presence of another loop, figure 4.39, is calculated in equation (4.11). The same figure shows on the right the principle behind preamp decoupling when using suitable low resistance low noise preamplifiers (LNAs).

$$V_{out} = V_{signal} + R_1 + i(\omega L_1 - \frac{1}{\omega C_1})I_1 + i\omega M_{12}I_2 \quad (4.11)$$

where M_{12} is the mutual coupling coefficient. The third term is addressed by tuning and matching, in effect leaving the real part, R_1 only. However, the fourth term

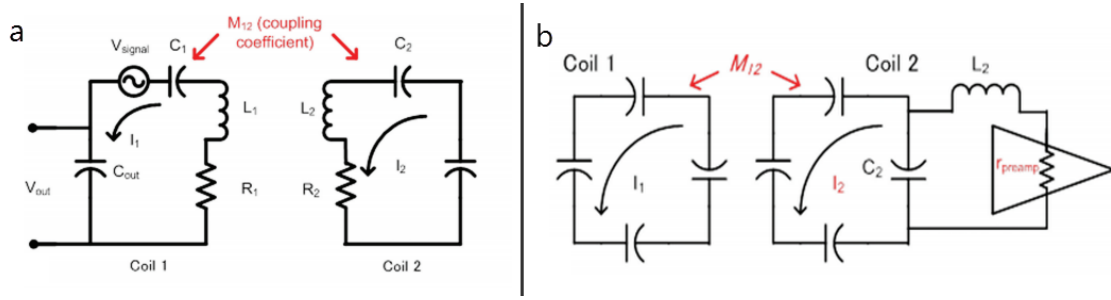


Figure 4.39: Schematics underlining the basic principle behind preamp decoupling. The free coils on the left are coupled by M_{12} which appears as noise proportional to $M_{12}I_2$ in the output of coil 1. The term is eliminated by adding the high impedance of the low preamp resistance, L_2 in parallel with the tuning capacitance C_2 . Reproduced from [160] with permission.

is the one we have been trying to eliminate so far. It can be minimised by either minimising the value of M which is in fact done by careful overlapping of the two coils, or by eliminating the current I_2 . Figure 4.39-b shows the principle behind preamp decoupling whereby a resonating LC circuit (r_{preamp} , C_2 and L_2) is inserted in series with coil 2. A careful selection of the components for that stage could make it resonant at the Larmor frequency. That will add a very high impedance in series to the coil, suppressing the current I_2 and zeroing the coupling term in the above equation.

The method is summarised in figure 4.40 where each of the receive loops is individually processed. The element is loaded and connected to an LNA with known input impedance. The latter is dependent on the specifications of the component and has usually a very low resistive part and a small inductive part. The LNA used here had an S_{11} of $0.95\angle 48^\circ$ and that was utilised to bring the impedance to an open circuit with the use of the appropriate coaxial cable length. The performance of the isolation is measured by the difference between the peak of the tuned coil in an S_{12} measurement with a double probe and the suppression of the same peak when the preamp is attached. The dip from each of the elements is shown in the network analyser photo in figure 4.41.

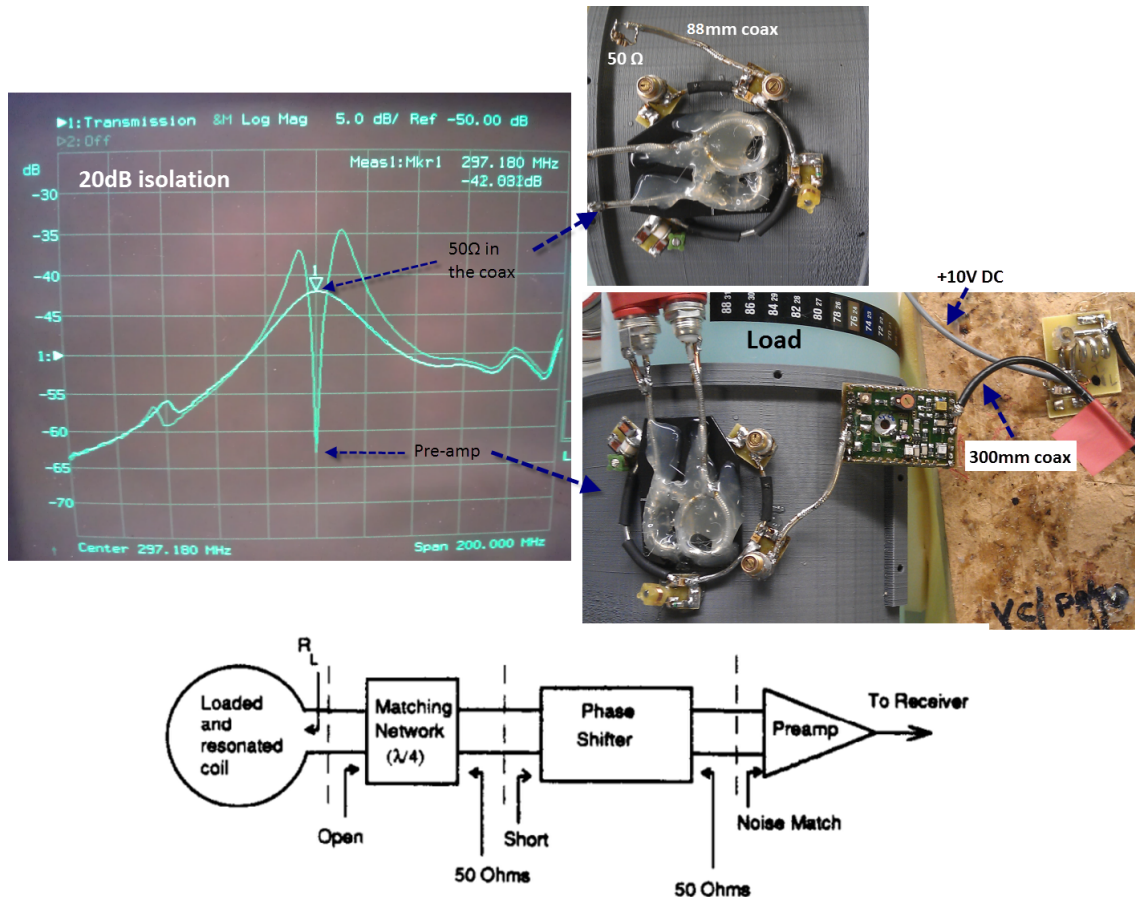


Figure 4.40: Process of passive decoupling with the use of low noise pre-amplifiers. Each element is adjusted individually. Initially tuned with the use of a 50Ω load. The network analyser photo shows the resonant peak for the 50Ω load with no decoupling and the dip with two peaks on the top shows the same measurement when the preamp is connected. The isolation was found to be 20 dB. The length of the coax cable is adjusted so that it transforms the non purely resistive impedance of the supplied preamp to a short circuit which is then transformed to an open circuit in series with the coil by the $\lambda/4$ matching network. Bottom image shows a generic approach to preamp decoupling. Reproduced from [161].

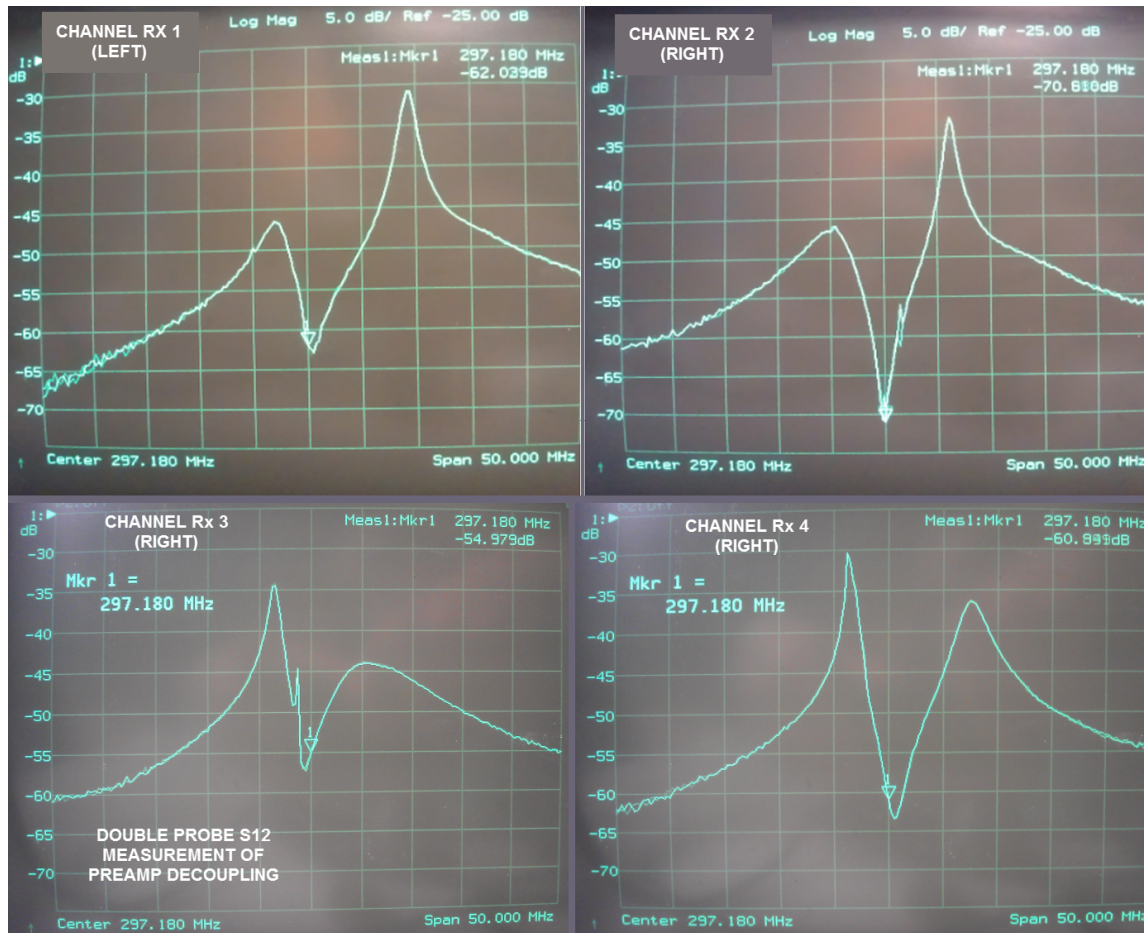


Figure 4.41: Passive decoupling dip and two peaks for the S_{12} double probe measurement of the receive channels. This graph shows that the loops exhibit a high impedance at the Larmor frequency suppressing the currents that would otherwise circulate in the loop. The isolation in each case is more than 20 dB. Preamp decoupling eliminates the mutual coupling on top of geometrical decoupling and increases the SoS SNR.

4.4.3 Cable Preparation

A critical step in coil construction is the cable preparation. As the coil design becomes more complex with multiple transmit channels and a separate receive array the RF coaxial and DC cables could interact with each other. If not taken care of, the interactions can be a major source of frustration and image quality degradation.

Preamp Oscillations

Preamp oscillation is a phenomenon whereby the output of the preamp is coupled with its input and the signal that is fed back is re-amplified, saturating the preamp, see figure 4.42. The system becomes unstable and adds noise to the receive chain. It is hence useful to minimise coupling around the Larmor frequency by suppressing the currents in the outer shield of the coaxial cable that leads from the pre-amp output to the scanner receiver components. For this reason shield traps are included as shown in the schematic in figure 4.36 and photo in figure 4.37 in both input and output leads of the preamps. The DC cables could be the carriers of the RF from the input to the output so the measures taken to avoid this are discussed next.

Bias Cable Decoupling

The DC cables are modified here in such a way as to eliminate interaction, with the first addition being the RF chokes, inductances at $2.2\mu H$ in effect blocking all the RF towards the DC source and preventing damage. That is especially important for the detuning circuitry in the Tx loops. In the same fashion, DC blocking capacitors are inserted after the diode and before the coil in order to avoid any DC currents in the main loop branches that could lead to perturbation of the RF field. The latter measure solves only half the possible problems though; the DC cables are placed near the RF cables, capacitors, the loops and preamps, and can become an RF power transmission platform from one stage to another. An adverse example can be preamp oscillation due to coupling between the output and the input as mentioned

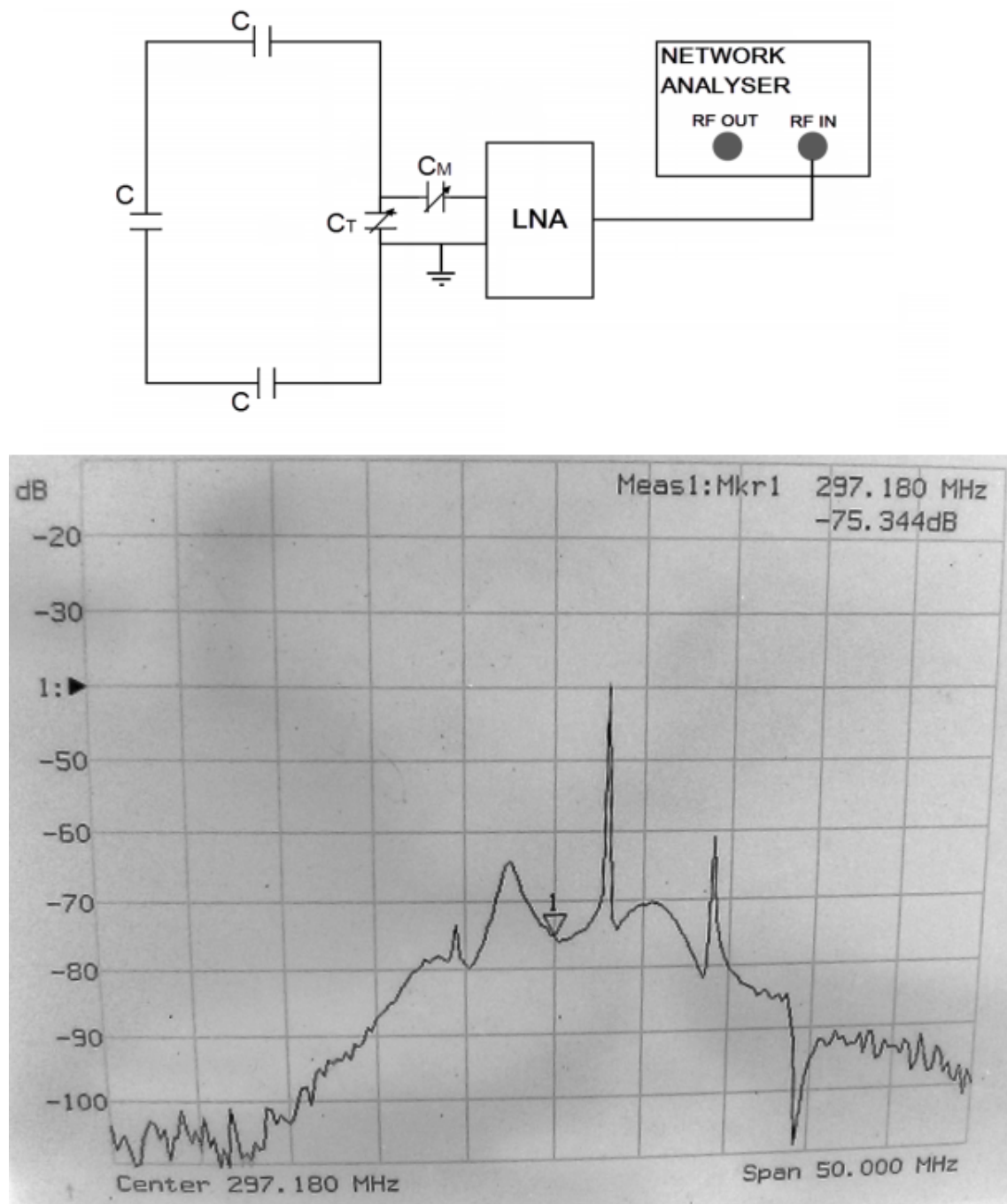


Figure 4.42: Example of preamp oscillations. The measurement connection diagram is shown on the top image. An S_{12} measurement is taken when the transmit port is left unconnected and the receive port is connected directly to the operating LNA. If the preamp oscillates, i.e. strong coupling between output and input, high peaks will appear. With the use of shield traps in the output and input, as well as DC cable preparation, the effect can be reduced.

previously.

One solution to the problem of DC cable interference is the introduction of common current suppression techniques as described for the RF cables. The option that was chosen for practical reasons in the construction of the neck coil was twisted pairs with occasional large capacitors between the live and the ground. Large capacitors, 1 nF, form a DC open circuit and an RF short, in effect connecting any RF common mode currents induced to the ground and leaving DC operation unaffected. The result is to reduce currents that are responsible for interference. See photo in figure 4.43 for an overview of the cable preparation in the coil.

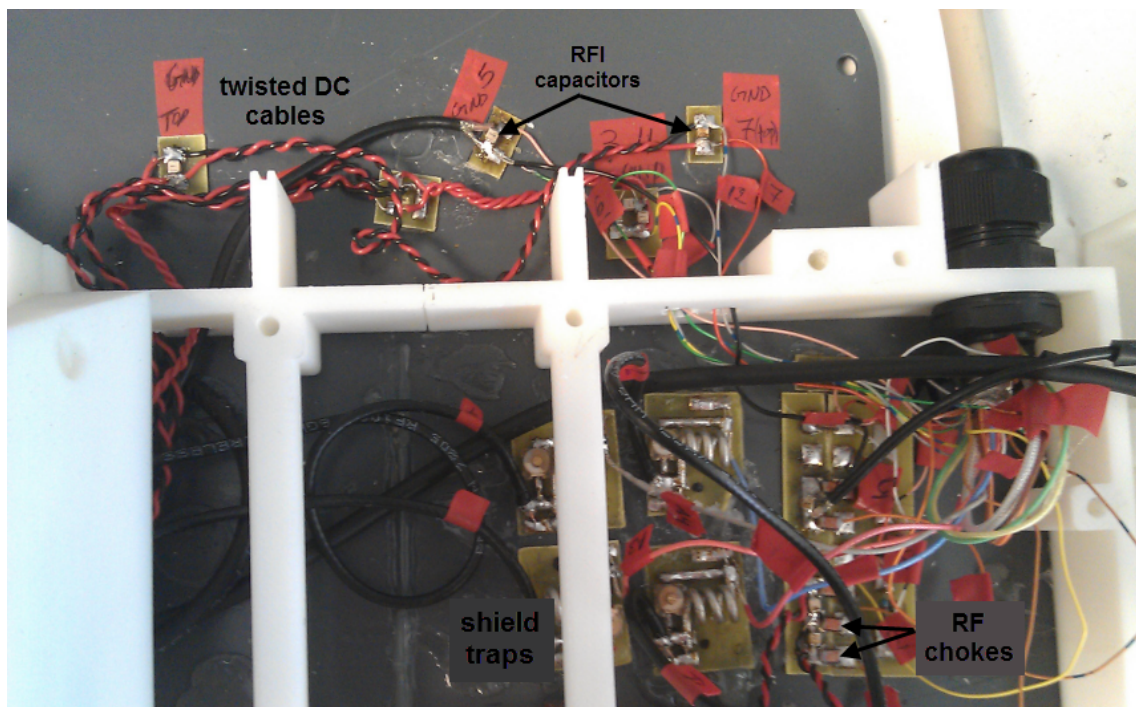


Figure 4.43: Photo of the coil base showing the DC cable measures taken to reduce RF interference. The wires are twisted so that there are no closed surfaces that can pick-up the stray RF from the Tx array. In addition, the radio frequency interference capacitors (RFI) direct anything induced to the ground. The RF chokes introduce a high impedance in RF band, hence protecting the DC sources from the high power RF pulses. The shield traps in the receive arrays suppress currents that would otherwise be induced from the coupling between the coax shield and the Tx array during transmission.

4.5 Summary

A 4 channel transmit / 4 channel receive coil has been constructed to operate at 7 T for black blood imaging of the carotids with the use of a DANTE preparation module. Based on electromagnetic simulations and a detailed account of the theory a coil was designed and constructed. It is expected to achieve adequate blood suppression and satisfactory resolution (that is SNR related) at an average location of the bifurcations of the carotid arteries.

In order to perform preliminary human trials, a number of standards have to be met as far as patient safety is concerned. The next chapter will concentrate on the evaluation of the coil safety and will attempt to determine the heating profile of the coil. Once power limits are set, the coil can be further evaluated with respect to what is required as discussed above. Simulations will play a dominant role in determining its safety and for that reason the accuracy and reliability of the simulations will need to be robustly scrutinised. Thus, an effort will be made to validate the simulations with experimental measurements.

Chapter 5

Safety Considerations & Evaluation

5.1 Introduction

While the magnetic component of the RF probe is the tipping ‘force’ on the precessing nuclei, there is an electric field component associated with it that is present at all times. In the previous chapter the design process and methods were described for how to limit the transmitted electric field in RF coils with the 4Tx/4Rx coil having been designed based on this approach. Regardless, the $\mathbf{E}$ -field present needs to be evaluated for its known effect on biological tissue.

One effective method to deal with the documented $\mathbf{B}_1$ inhomogeneity at ultra high fields (7 Tesla and above) is the already discussed method of parallel transmission (pTx) [127, 162]. Since pTx requires the use of more than one transmitting source, RF safety management becomes more complicated. In addition, large volume body coils are not practical at these field strengths and only local surface coils are used, adding an extra risk from the increased localised power deposition that this type of coil exhibits.

With regards to pTx, the RF pulses from multiple surface coils are adjusted

in such a way that the superposition or cancellation of the waves improves the field uniformity in the region of interest (ROI). Parallel transmission can involve the alteration of the profiles of the RF pulses to achieve the desired excitation but the most common method is to manipulate only the magnitude and phase whilst the modulation envelope transmitted in each channel is kept constant. The latter methodology of pTx is called RF shimming [163]. However, the benefit from the B_1 mitigation in a given ROI can turn out to be a hazard in another ROI area via the formation of local hot spots.

An additional risk arises in the case of a malfunction in the RF power amplifier. If the input phases and magnitudes are not kept below the error margin, unpredicted wave superposition can produce electric field peaks in the subject. In order to account for such an adverse occurrence, constant monitoring of the transmitted waves is required by the system. Thus, more specialised infrastructure is needed by scanners that employ pTx in order to adequately supervise the power deposition. A detailed account of the safety supervision system of the scanner used for the experiments in this thesis is given in section 5.3.

Simulations and phantom measurements are the most accepted methodologies to estimate and to secure safety during pTx scans [164, 165]. For instance, a conservative approach is to add the electric field magnitudes generated by each transmit element in simulations of anatomically realistic models and to restrict the transmission based on the maximum possible estimated local power absorption. The latter would lead to very strict power limits that could be an unsurpassable obstacle in obtaining the required flip angles in the ROI.

An RF burn is a valid concern and needs to be taken into account at all times and in this chapter the safety aspects of the constructed 4Tx/4Rx coil is outlined in detail. The approach can vary from conservative to realistic imposing, respectively, conservative or realistic power limits. In particular, a novel methodology of RF heating supervision is adopted, developed from Eichfelder et al. [166]. The Virtual

Observation Points allow the scanner to calculate the power absorption during a scan, thus maintaining safety and improving the power efficiency of a coil.

5.2 RF Safety

The result of a current density in biological tissues induced by an alternating electric field is microscopically an increase in the internal energy of the water molecules and macroscopically an increase in the temperature. However, the relation of the temperature of a living organism versus the deposited power in its tissues is far from linear and is rather complex. In fact, the human body is able to keep its temperature within a fairly well defined range for a wide variation of external conditions, such as environmental temperature, humidity or electromagnetic power deposition.

The mechanism behind the ability of an organism to keep its body temperature constant is called thermoregulation. It is complex and beyond of the scope of this thesis to investigate how the tissues in living organisms keep their temperature within a particular range and how they compensate for external thermal stimuli. This thesis will concentrate on the effect that the RF heating has on tissues without taking into account internal parameters of thermoregulation and will assume standard environmental conditions (such as room temperature).

Normal human body temperature is approximately 37°C and can vary on occasions from 35°C to 42°C; anything below or above poses a risk to the life of the person. Hence the most reliable and unchallenged limits that one can set when it comes to RF safety is the effect it has on the temperature of the subject. Any MR medical device (either RF or not) is regulated by the International Electrotechnical Commission (IEC) in its 60601 documentation concerning medical electrical equipment, [167]. The table in figure 5.1 summarises the maximum allowed absolute core and local tissue temperatures for different control levels.

First level Controlled Operating mode is when the operation of the MR equip-

Operating mode ↓	Maximum CORE TEMPERATURE	Maximum Local tissue temperature	Rise of CORE TEMPERATURE °C
NORMAL	39	39	0,5
FIRST LEVEL CONTROLLED	40	40	1
SECOND LEVEL CONTROLLED	>40	>40	>1

Figure 5.1: Temperature limits set by IEC in the 60601 documentation. For first level controlled operation mode, the maximum local temperature can be up to 40°C . Only volume coils will likely have an effect on the core temperature whilst local transmitters are more likely to affect the temperature in localised areas. Reproduced from [167].

ment generates one or more outputs that can reach a value that can cause physiological stress to patients which need to be controlled by medical supervision. This is the mode under which pTx transmit will be assumed to operate and all the power limits will be subject to the specifications set for this mode.

5.2.1 Specific Absorption Rate

The result of an electric field, $\mathbf{E}$, in a conductive material is current density, $\mathbf{J}$, as dictated by Ohm's law, $J = \sigma E$, σ being the conductivity of the material. $\mathbf{J}$ in turn produces heat at a rate equal to $P = |J|^2/\sigma$ which can be re-written as $P = |E|^2\sigma$. If the mass of the conductive material is taken into account then a new quantity can be derived, the *Specific Absorption Rate* or SAR. SAR is the variable that quantifies the RF power deposited per unit mass and is mathematically defined in equation (5.1).

$$SAR = \frac{|E|^2\sigma}{2\rho} \quad (5.1)$$

SAR is measured in units of Wkg^{-1} where σ is the electrical conductivity in Sm^{-1} , $|E|$ is the peak amplitude of the electric field in Vm^{-1} and ρ is the mass density in kgm^{-3} .

As discussed previously there are two types of coils, volume and local. The names

denote their effect on the subject with the volume coils in general depositing energy over a large portion of the body and the local coils over a smaller, localised area. In order to regulate the power deposition and RF heating, three types of SAR have been defined, and some are more or less relevant to whatever type of coil is used.

The table in figure 5.2 summarises the SAR limits set for volume transmit coils. Two types of SAR are presented here, whole body SAR and partial body SAR, the latter being known also as global SAR. Whole body SAR is simply all the power transmitted over the total patient mass, which would typically exceed 60 kg. This type of SAR refers to large body size coils with the most typical example to be the embedded birdcage coil installed on clinical scanners. These coils have the capacity to raise the core body temperature but their SAR supervision is straightforward at 3T or less.

The partial body SAR is also applicable to body coils but also to smaller transmitters such as local coils. Its calculation is derived from the power transmitted from a coil over the *exposed patient mass*. The latter is defined as the patient mass in the *effective volume* of the transmit coil which is the volume in which 95% of the total absorbed power is deposited inside a homogeneous dielectric material filling the patient space. It is obvious that this is slightly more complicated than the whole body SAR since an estimation of these parameters needs to be made. This type of SAR needs to be conservatively calculated or supervised when local and volume transmit coils are in operation.

The third unmentioned type of SAR is the local SAR that is specific to local transmit coils. Since its consideration and calculation is the most complex of the three types of SAR, it is discussed in more detail below.

5.2.2 Local SAR Limits

Following the definition of the two previous SAR types, the table in figure 5.3 summarises the limits set for local or surface transmit arrays like the 4Tx/4Rx coil.

Averaging time	6 min		
	WHOLE BODY SAR	PARTIAL BODY SAR	HEAD SAR
Body region →	Whole body	Exposed body part	Head
Operating mode ↓	(W/kg)	(W/kg)	(W/kg)
NORMAL	2	2 – 10 ^a	3,2
FIRST LEVEL CONTROLLED	4	4 – 10 ^a	3,2
SECOND LEVEL CONTROLLED	>4	>(4 – 10) ^a	>3,2
Long MR EXAMINATION specific absorbed energy	The maximum allowed specific absorbed energy is 14,4 kJ/kg (= 240 W·min/kg) per MR EXAMINATION provided that the limits of this standard are still met		
Short duration SAR	<u>The SAR limits over any 10 s period shall not exceed two times the stated values.</u>		

^a The limit scales dynamically with the ratio "exposed PATIENT mass / PATIENT mass":

NORMAL OPERATING MODE :

PARTIAL BODY SAR = 10 W/kg - (8 W/kg * exposed PATIENT mass / PATIENT mass)

FIRST LEVEL CONTROLLED OPERATING MODE :

PARTIAL BODY SAR = 10 W/kg - (6 W/kg * exposed PATIENT mass / PATIENT mass)

Figure 5.2: SAR limits for volume transmitters. The limits relevant to this study are marked in red. For first level controlled use the allowed head SAR over a 6 minute period is 3.2 W/kg and twice this value for any 10 second duration. Reproduced from [167].

Surface coils are positioned very close to or against the skin of the subject and thus they become very ‘power to SAR’ efficient which means that for a small power input the power deposited in the vicinity of the coil is very high per unit mass. These coils cannot generally interfere with the body core temperature but can cause, in adverse conditions, excessive localised heating and tissue damage with 1st and 2nd degree burns being documented in the past [168].

Apart from the allowed power per unit mass there is a limitation of a 1°C increase over a 6 minute rate for the head. This limitation adds a total energy allowance for different tissues. From equation 5.11 discussed later in this chapter, it is clear that even for the same SAR in two particular tissue types the temperature increase over the same duration depends on their specific heat capacity. Nevertheless, the SAR is a conventional quantity for RF engineers to monitor, and its limits are set based on the most sensitive tissues, i.e. the ones that will rise by 1°C fastest.

The calculation of local SAR involves 10 gram tissue mass averaging of the ex-

Averaging time	6 min		
	LOCAL SAR		
Body region →	Head	Trunk	Extremities
Operating mode ↓	(W/kg)	(W/kg)	(W/kg)
NORMAL	10 ^a	10	20
FIRST LEVEL CONTROLLED	20 ^a	20	40
SECOND LEVEL CONTROLLED	>20 ^a	>20	>40
Short duration SAR	<u>The SAR limits over any 10 s period shall not exceed two times the stated values</u>		
^a NOTE In cases where the orbit is in the field of a small LOCAL RF TRANSMIT COIL, care should be taken to ensure that the temperature rise is limited to 1 °C.			

Figure 5.3: Limits of local SAR for a 6 minute scan. In the first level controlled operation mode the limits are set at 20 W/kg for the head with no more than 40 W/kg for any 10 sec period. In addition, any temperature rise due to RF in the scanned subject should be no more than 1°C. Reproduced from [167].

posed patient mass. That is a process whereby the exposed patient mass is separated in to cubes, spheres or any regular shapes with each one containing 10 grams of tissue. The local SAR estimation has thus far been only an ‘off-line’ calculation due to the fact that the **E**-fields cannot be reconstructed during a scan. In particular, there is no way to acquire a conductivity/density map for a scanned subject and even if it were possible, the computational time would be too long. Thus, electromagnetic simulations, i.e. Maxwell equation solving algorithms, are used for estimation of **E**-fields over anatomically realistic human phantoms. The accuracy of the prediction depends on the use of accurate simulation conditions and if available multiple human phantoms covering a range of shapes and body masses.

The local SAR becomes more crucial when it comes to surface arrays like the 4Tx/4Rx coil. As discussed earlier, the purpose of the 4 separate transmit channels is to mitigate the **B**₁ field in the ROI. However, 4 inhomogeneous **E**-fields are superimposed increasing the chance of creating ‘hot spots’ somewhere in the exposed patient mass. Also, the fact that there are multiple surface coils surrounding a body area gives a volume transmit dimension that needs to be addressed by supervising the global SAR. Thus, when transmitting with one surface coil monitoring of the local SAR is adequate but when transmitting with an array coil both local and global

SAR need to be taken into account.

The worst case approximation is a common way to ensure that both SARs stay below their respective maximum allowable [169, 170]. For the global SAR this is the total allowed power transmitted by any of the channels individually or in combination that would reach the 3.2 W/kg limit over the exposed patient mass. The worst case local SAR would be calculated by estimating the maximum allowable power per channel required for a particular phase combination to reach 20 W/kg within any 10g mass contained in the subject. However, the methodology to be used for local SAR depends on the power monitoring systems installed in the scanner. For this reason, a detailed account of the RF supervision components of the scanner used in this thesis is given in the next section and the issue will be re-addressed later.

Each transmit source, feeding either a surface or a volume coil, will generate a complex $\mathbf{E}$ -field that is represented by a magnitude and phase when a zero phase source is transmitting, see figure 5.4 and 5.5 for the $\mathbf{E}$ -field of the side Tx element of the 4Tx/4Rx. The fields from the same coil on a human phantom can interact in a complex way and it is challenging to comprehend them intuitively, see figure 5.6. These calculated individual $\mathbf{E}$ -fields are then used to calculate the local SAR maps based on tissue conductivity and density.

Averaging over 10g tissue masses can be regarded as the lowest allowed resolution for local SAR and a finer one that entails better estimation is generally desirable. However, very high resolutions, i.e. sub millimetre, become irrelevant with respect to probable phase and magnitude error and divergence with the real scan. To illustrate how the averaging affects the resolution and accuracy of SAR prediction four different resolutions are presented all based on an isotropic 1 mm voxel simulation of a male phantom, see figures 5.7 and 5.8. More about the averaging resolution that will be used as well as the specifications of the simulation will be found in section 5.4.2.

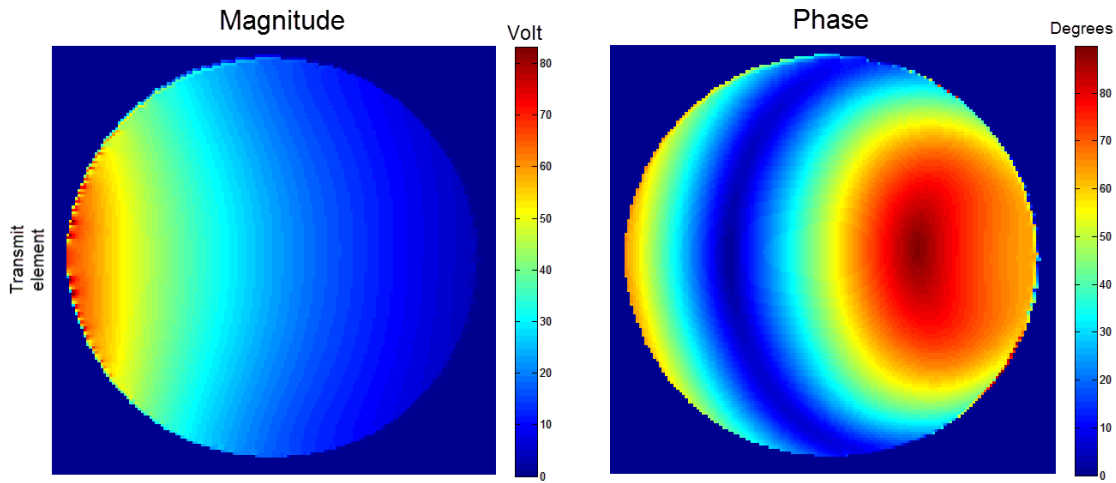


Figure 5.4: Simulations showing the side element transmitting against the surface of a homogeneous cylindrical phantom. The magnitude, on the left, decays from about 80 V/m close to the element to below 10 V/m on the other side. The phase indicated in the right image shows how the $\mathbf{E}$ -field varies in the dielectric material. With regards to the phase, a sinusoidal pattern can be observed due to the homogeneity of the particular phantom.

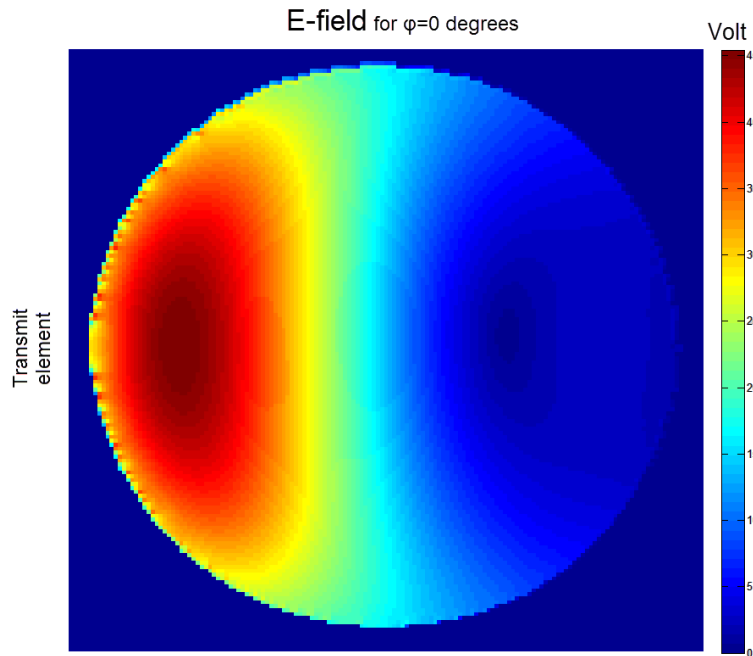


Figure 5.5: The E field ‘seen’ by the dielectric material for any phase is $E = |E| \cdot \cos(\omega t + \theta(x, y))$ where $\theta(x, y)$ is the phase in figure 5.4. The field map is shown for $t=0$ and hence $\phi = \omega t = 0$.

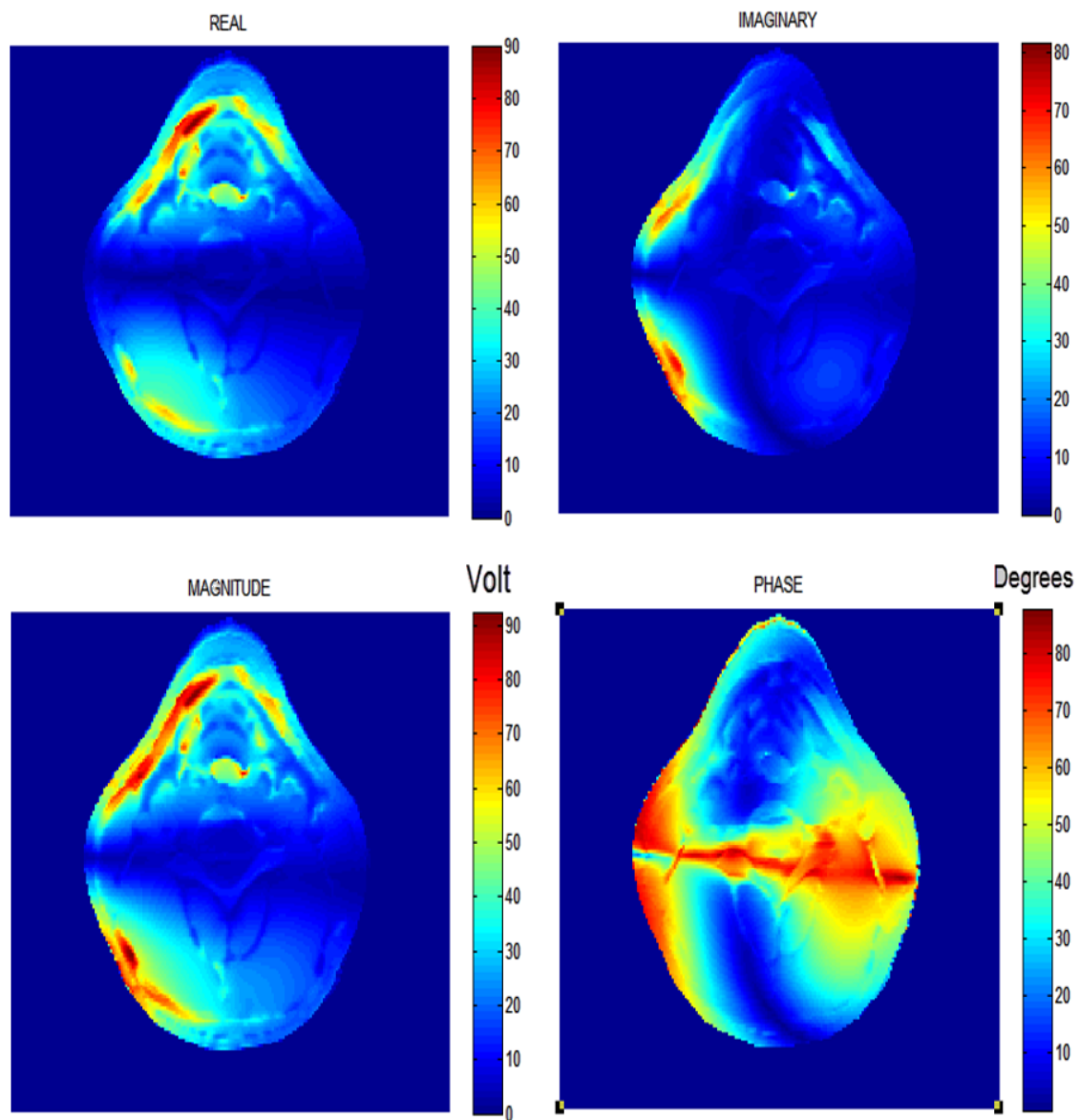


Figure 5.6: The $\mathbf{E}$ -field variation along a computational phantom. On top the real and imaginary matrices can be seen and on the bottom row the magnitude and phase. Intuitively it is very challenging to understand what the real and imaginary components mean and thus the magnitude and phase is used for observation, but calculations are done only via the former structure.

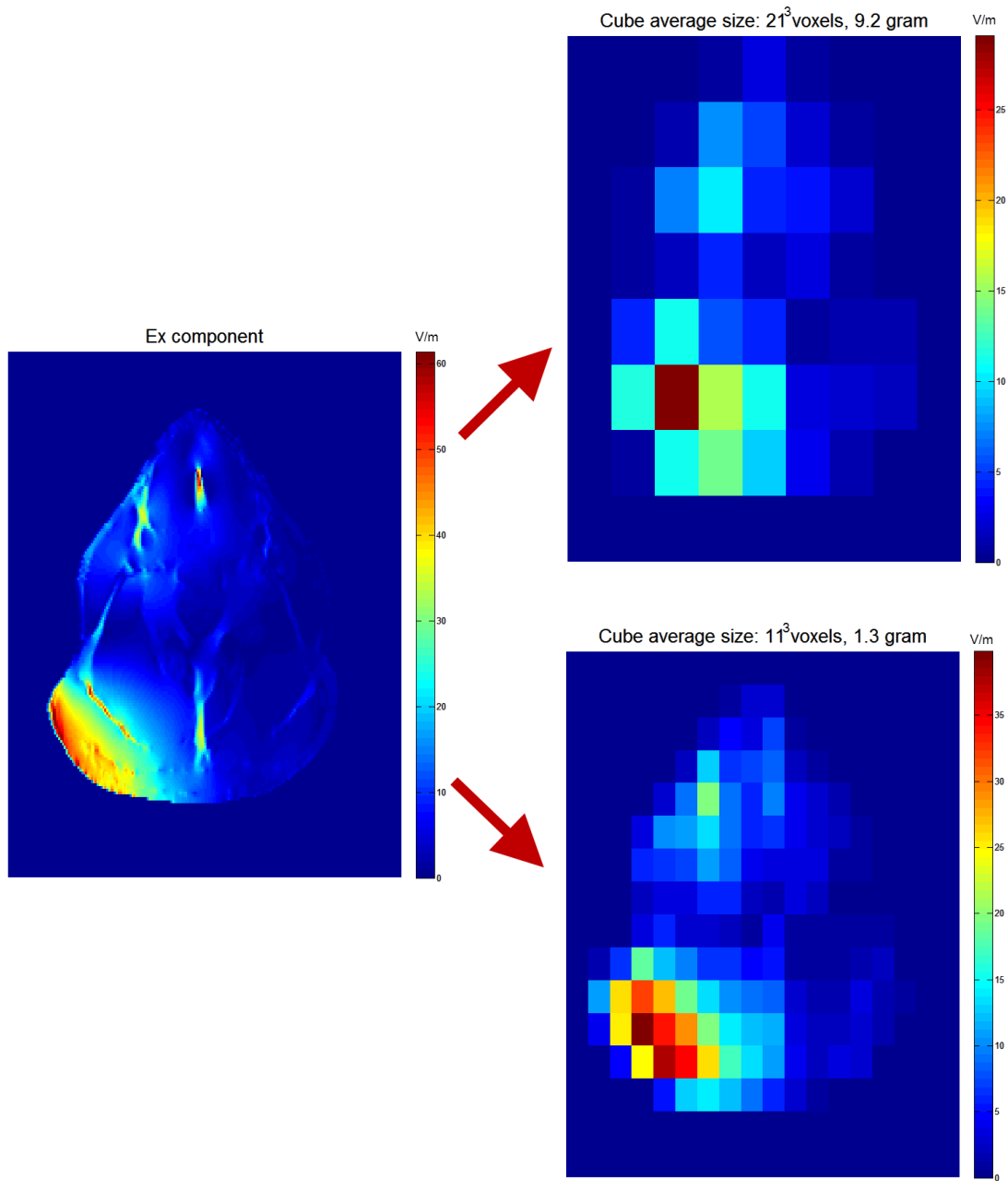


Figure 5.7: Comparison of different averaging resolutions for the x $\mathbf{E}$ -field component, shown on the left column (top and bottom) at 1 mm isotropic resolution. On top a 9.2 gram/voxel estimation that is the best 10g approximation is presented. The resolution is very limited with about 30 active voxels per slice. Clearly any fine detail is missed with this approach but it remains an attractive method in that it produces the fewest SAR monitoring variables. On the bottom row, a 1.3 gram per voxel estimation is presented, that is an approximation of the, also standard, 1g averaging that shows the resolution improvement.

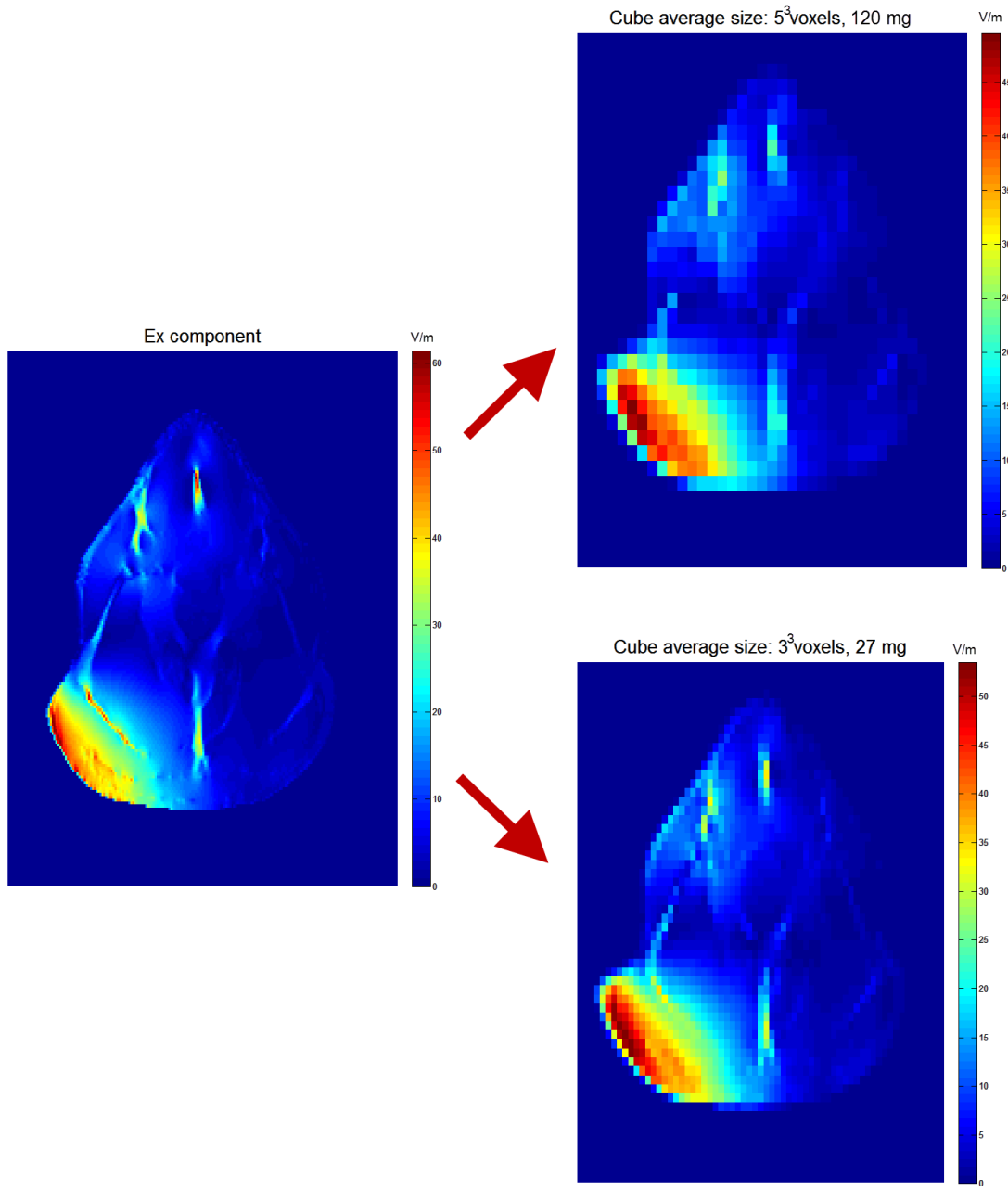


Figure 5.8: Higher resolutions are shown here that are not required by the IEC but are presented and used here for their improved resolution and reliable reconstruction of the calculated $\mathbf{E}$ -field map. A full grid, i.e. 1mm isotropic, can be regarded as unnecessary due to the expected variability between simulation and experiment that arise due to the subject variability, phase errors and hardware effects. Averaging thus is a compromise between the error margin and the accuracy of the estimation of the RF heating under a particular transmitting scheme. Since handling multiple 1 mm isotropic matrices proved computationally demanding, a 27mg averaging will be performed on the $\mathbf{E}$ -fields to be used for the calculation of SAR in this study. The benefit from such averaging is the reduction of the raw data matrix by one order of magnitude and hence a significant improvement in processing times.

In the next section, 5.4, the different SAR quantities will be calculated based up on the available supervision systems detailed in section 5.3. The particulars of the SAR approximation based on simulations as well as the simulation validation with experimental measurements will then follow.

5.2.3 Other Hazards

Even though some other hazards will not be addressed here, it is useful to mention briefly that, apart from RF coils leading to RF burns, other equipment, such as physiologic monitoring wires, external accessories or objects that contain conductive materials can and have caused excessive heating in the past inflicting burn injuries to patients undergoing MRI [168]. Implants are a major concern not only for heating but also for malfunctioning due to the strong static field, and therefore can pose a grave danger to the patient. The primary problem during an MRI scan for such devices can be the creation of conductive loops in the patient. This is realised either by the monitoring wires themselves or by conductive loops within the human body. For this reason consideration must be given to the arrangement of the wiring and the patient posture in the table.

Many incidents of excessive heating have been reported in patients undergoing MR procedures that were unrelated to equipment problems or the presence of conductive external or internal implants or materials [168]. These incidents included first, second and third degree burns that were experienced by patients. In many of these cases the reports indicated that the limbs or other body parts of the patients were in direct contact with the RF coils or other transmitting elements. In other cases, skin to skin contact points were suspected to be responsible for these injuries. At the skin point the resistance is higher which results in heat deposition from the induced currents in the body loop.

In general, these occurrences are avoidable, have been documented and safety guidelines are clear and unchallenged. Consequently, there is no need for further

investigation and these risks are not of particular interest in this thesis.

5.3 System Description

It is useful at this point, before the coil validation, to describe the pathway of RF monitoring in the scanner to which the coils will be interfaced. Referring to the diagram in figure 5.9 the process starts with the sequence that is loaded in the scanner. It is optimised and sent out to the Measurement Control Units (MPCUs), one for each of the 8 transmit channels, where the RF waveform is shaped to be the input of the RF Power Amplifier (RFPA). At the output of the RFPA the first monitoring system already performs a check, and is the Transmit Antenna Level Sensor (TALES). The latter comprises of 8 broadband phase insensitive power meters that measure the total power from 10 to 300 MHz and the output is processed by the Power Absorption Limiter Multi-Channel (PALI MC). The PALI MC in its turn can abort the scan if limits set by the operator are exceeded. Power limits can be set for each channel for various time integrations but most commonly total power limits are set by the operator based on SAR estimations as discussed in section 5.2.1.

After the TALES and the PALI MC the power is in effect directed to the coil plug where the coil is connected. However, Directional Couplers (DiCo) are placed closer to the plugs and pick up a small portion of the power in order to detect the magnitude and phase of each of the 8 channels. The DiCo data are directed through a switch box, similar to a TR switch but with inverse operation (during Tx the DiCo output is directed to the receiver), to the MR receiver that records them with an Analog to Digital converter (ADC), the same one that is used for standard MRI signal reception. Once the DiCo data have been sampled they are sent to the MR imager that in effect is the image reconstruction computer. There, local SAR calculation takes place via pre-calculated E-field maps in anatomically realistic models. The method of Virtual Observation Points (VOPs) used in the scanner is

one averaged measurement for successive time frames but the average is updated every $10\mu s$ sample. The sliding window averaging can then provide the minimum reaction time in the event of a power transmission limit breach. The TALEs accuracy is 12% and since the physical position of the system is proximal to the RFPA there is a nominal 1.9dB attenuation from its output to the coil plug due to the cabling. Accuracy must be taken into account when setting power limits but cable attenuation can be disregarded for conservative supervision. Information about the TALEs timing and limits is summarised in the table in figure 5.10.

Averaging time period	Limit	Error trace when limit is exceeded
10 μs	1400 W per channel forward power Relevant for component protection	Limit violation: Peak active forward
10 ms	Limit = 8000 W Sum of RF power forward - power reflected	Limit violation: 10 ms integral fwd - ref
10 s	Limit user defined Sum of power forward - power reflected	Limit violation: 10 sec integral fwd - ref Refers to sum-channel
6 min	Limit user defined Sum of power forward - power reflected	Limit violation: 6 min integral fwd - ref Refers to sum-channel
Unexpected RF	Limit = 3 W Sum of RF power (only sum-channel is supervised)	Limit violation: Peak unexpected forward Refers to sum-channel

Figure 5.10: TALEs power measurement specifications. Sampling rate of the system is at 100KHz, i.e. every $10\mu s$. Any peak above 1400 W per channel is deemed excessive for the components and PALI MC stops the scan immediately. At 10ms averaging step the limit is set at 8000 Watt, i.e. the total power available at the RFPA in combined mode. At 10 sec and 6 min the limit is set by the user as defined in the IEC60601 documentation. TALEs also accounts for unexpected RF that can indicate a system fault. Taken from [171].

The use of VOPs covers the supervision of local SAR based on a pre-determined electromagnetic model of the subject, while global SAR limits are still set by the operator based on the exposed patient mass. In the absence of VOPs the supervision needs to be based on the worst case scenario of either the local or global SAR.

Whichever is the worst will define the limits that need to be set at 10 sec and 6 min.

RF safety parameters and the supervision system of the scanner are now both described making it possible to proceed to the safety evaluation of the constructed 4Tx/4Rx neck coil in the next section.

5.4 4Tx/4Rx Coil Safety Validation

The three basic steps in safety validation of an RF coil can be broken down to:

- Electromagnetic Simulations with realistic anatomical models.
- Estimation of the Worst Case SAR via extracted **E**-fields.
- Experimental Validation of the Electromagnetic Simulations.
- Use of the coil under the particular power limits.

In this section, the first two steps will be described with the outcome being the definition of the power limits. The crucial step of simulation validation will be described in the subsequent section.

5.4.1 Model & Construction

A specifically designed coil was constructed as discussed in the previous chapter and modelled to fit the neck, see figure 5.11. A passive decoupling scheme, along with active detuning, was adopted for the receiver loops in order to secure decoupling during transmission. The schematic of the receiver loop is also included and discussed in chapter 4. The isolation with active detuning and a power input low enough to keep the crossed diodes inactive was measured at the highest value of -16.6 dB for Tx1 to Rx1 with an average value of -32.3 dB. The activation of the crossed diodes brought a further -5 dB isolation in the worst case. For a full account of the isolation see table 5.1.

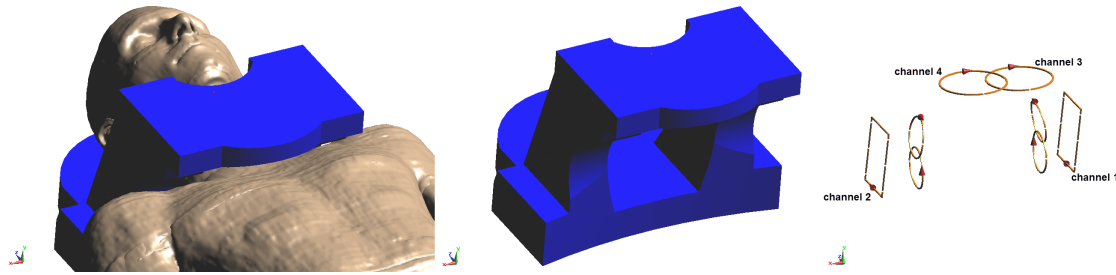


Figure 5.11: The transmit-only receive-only (TORO) neck coil was constructed based on the design considerations as presented in chapter 4. The coil consists of a 4 channel transmit array enclosing a 4 channel receive array.

	Rx1	Rx2	Rx3	Rx4
Tx1	-16.6dB	-17.2dB	-36.4dB	-41.7dB
Tx2	-40.3dB	-53dB	-25.3dB	-22dB
Tx3	-25.3dB	-24.3dB	-36.2dB	-37.8dB
Tx4	-37.5dB	-33.6dB	-35.7dB	-34.2dB

Table 5.1: Isolation between transmit and receive arrays during transmission mode. The crossed diodes are not activated, upon activation they would provide about 5dB additional isolation. More thorough analysis on the scattering parameters can be found in Chapter 6.

For the accurate estimation of SAR the phase added at each Tx channel was measured from the transmit plug. The results are shown in Table 5.2. The BALUNs were slightly different for channels 1-2 and 3-4 in that the first had 3 windings of 5mm diameter and the latter had 3 windings of 7.3 mm diameter.

	Tx1	Tx2	Tx3	Tx4
Phase [deg]	-38.6°	-44.6°	27.8°	25°
Positive [deg]	6°	0°	72.4°	69.6°
Rad	0.10472	0.00000	1.26362	1.21475
Scan input	1.1589	1.2636	0.0000	0.0489

Table 5.2: Tx Phase relations, all in degrees apart from the last line which is in rad. The differences are due to different cable types used for the side and the top coils even though their length was selected to be similar.

5.4.2 Simulations & Averaging

Simulation Specifications

Simulation software was used for the estimation of the electric fields in two anatomically realistic models from the virtual family pack, a 32 year old male - Duke and a 27 year old female - Ella, [147, 172]. The Finite Difference Time Domain (FDTD) method is used in the package Semcad-X, Speag, Zurich to solve the Maxwell equations. The field sensor could be selected to have 1 mm isotropic voxels helping the calculation of mass averaging.

The Tx coil was simulated in the absence of the receiver array since the two are considered sufficiently decoupled during transmission. Initially, broadband simulations were run to tune the coils by re-adjusting the capacitance in multiple iterations but with fixed capacitors set at the same values as the real coil. The circuit schematics, the physical coil and the CAD model are all detailed in the previous chapter.

Mass Averaging

After the calculation of the $\mathbf{E}$ -fields, the SAR as defined in (5.1) can be calculated by either of the following options:

- Use of equation (5.1) and then averaging over SAR values.
- Averaging the $\mathbf{E}$ -field and then performing a SAR calculation.

The first option is simpler in terms of process because it does not entail averaging of complex matrices. However, depending on the requirements of the application, i.e. VOPs, the second option was deemed suitable. The VOP algorithm for local SAR supervision requires the averaged $\mathbf{E}$ -field components, E_x , E_y and E_z . These maps are complex and the averaging process is applied to each field from each transmit element. The averaging is shown mathematically in equation (5.2).

For any given averaging element mass, the appropriate number of voxels is selected and their average is calculated. If the whole matrix here is one elemental

tissue mass the average for n voxels is:

$$E_{avg}^1 = \frac{a_1 + a_2 + \dots + a_{n-1} + a_n}{n} + i \cdot \frac{b_1 + b_2 + \dots + b_{n-1} + b_n}{n} \quad (5.2)$$

The same applies to each separate **E**-field, E_l , generated by the l^{th} transmit loop. The averaging takes place before any superposition or combination of the fields.

The chosen averaging resolution is a factor that depends on available computational power (i.e. memory and GBU) for the required calculations. For this application the initial resolution is a 204 by 184 by 313 grid that corresponds to 1 mm sized cubical voxels of a male and female phantom as shown in figure 5.11. The selected resolution is the $3 \times 3 \times 3$ voxel cubes, i.e. $3\text{mm} \times 3\text{mm} \times 3\text{mm}$ cubes that for a 1000 kg/m^3 density gives 27 mg. The resolution is accurate enough to contain **E**-field variation details that are lost at lower resolutions, see again figures 5.7 and 5.8 in section 5.2.2.

Channel Phase Variation

Based on table 5.2 the phase differences have to be included for any reliable estimation of RF heating due to superimposed fields from different transmitting elements. The simulations are then adjusted for the phase differences for each source in the model. The four sources are transmitting as shown in figure 5.12. These values are compensated with opposite fixed time delays at the RFPA. The RF pulses will propagate through the cables and reach the probe in phase with each other. The directional couplers will sample the pulses before the coil plugs with the compensated values set by the operator. The local SAR calculations in the imager will be done with these phase differences and hence the local SAR is accurately estimated.

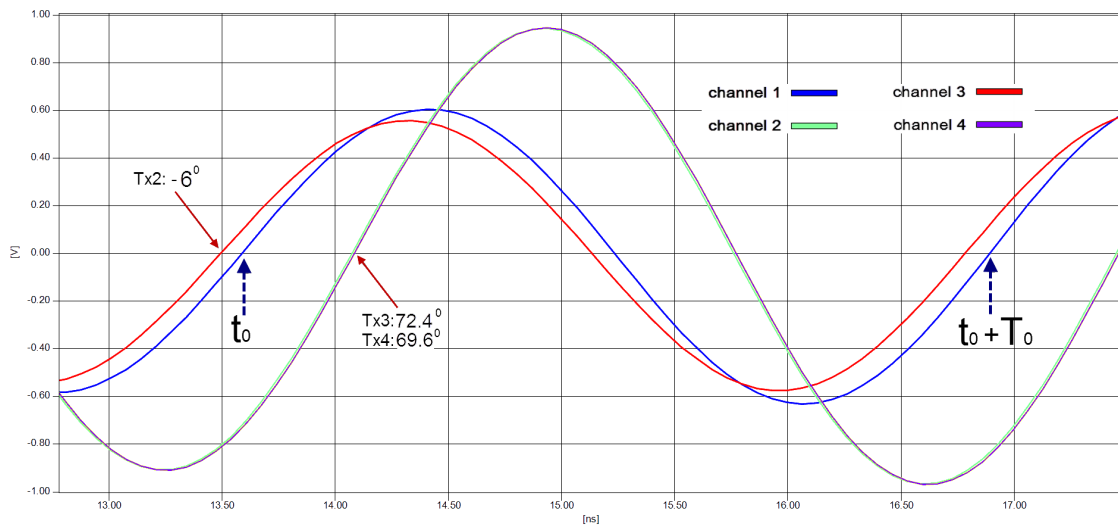


Figure 5.12: The source voltage time diagram is shown here for the four Tx elements during the simulation. The appropriate delay was selected for each source in order to match with the measured cable and BALUN variation for each element. The magnitude discrepancies are due to the model loop resistance but the extracted results are normalized for sources with 1W input.

Power Normalisation

The matching capacitor is not used in the simulation model due to the complexity of including such a stage in a simulation as well as the multiple steps that would be required to match the coil to the load, see figure 5.13. The software provides the option to exclude the mismatch and thus Z_{in} in figure 5.14 is set to be 50Ω without changing the coil resonance.

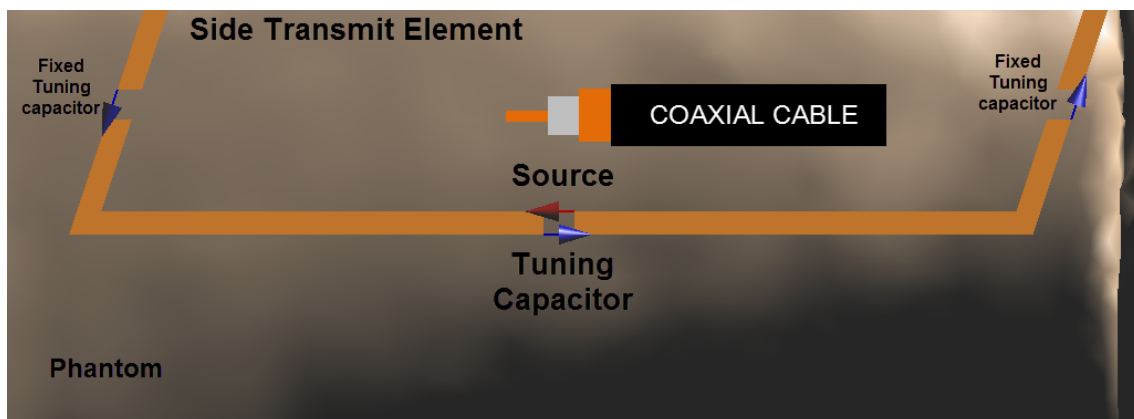
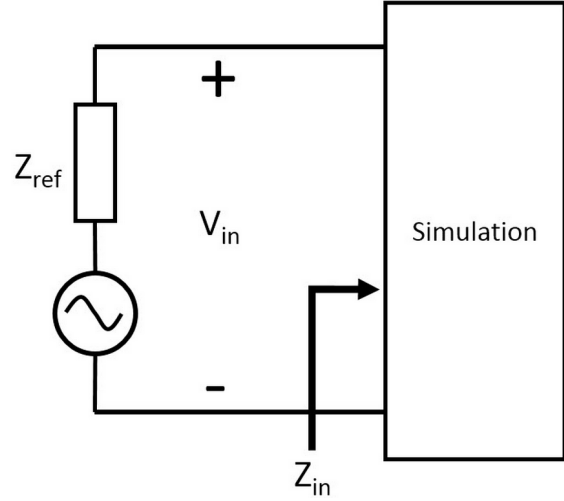


Figure 5.13: Semcad X model of the feeding stage. The source is not connected via a matching capacitor but the output electric field can correspond to a fully matched source with a selected driving voltage.

Figure 5.14: Z_{ref} is the reference resistance used for the system, 50Ω , and V_{in} is the voltage input driving the coil or simulation. In the physical coil the 50Ω source is the amplifier driving the coil through the 50Ω coaxial cable.



By normalising each of the $\mathbf{E}$ -fields, the driving voltage of each source in the simulation can be defined. The requirement of the monitoring algorithm is to use sources of 1 V rms for 50Ω impedance. That leads to a $P_{in}^{rms} = 0.02$ W. The ‘target normalisation power’, P_{norm} , is derived as in the equation (5.3) and is then applied before the extracted $\mathbf{E}$ -field.

$$\begin{aligned}
 P_{in} &= \frac{V_{in}^2}{R_s} \\
 \frac{P_{norm}}{P_{in}} &= \frac{V_{norm}^2/R_s}{V_{in}^2/R_s} \\
 P_{norm} &= P_{in} \left(\frac{V_{norm}}{V_{in}} \right)^2
 \end{aligned} \tag{5.3}$$

where P_{in} is the delivered power without matching, V_{in} is the voltage seen at the source, R_s is the source impedance, and P_{norm} is the target power normalisation. The post-processor of the software calculates what necessary amplification is required in the V_{norm} , i.e. the source voltage, so that the P_{norm} reaches its target. The source resistance is set at 50Ω .

For a P_{norm} rms value of 0.02 W and the mismatch disregarded, V_{in} is effectively 1 V rms. Hence, all the $\mathbf{E}$ -fields after calculation were normalised for 0.02 W plus 12% power error which is specific to the scanner supervision system. Hereafter,

all the SAR matrices and VOPs will have as a starting point the 27 mg averaged $\mathbf{E}$ -fields for zero phase sources of 1.12 V rms.

5.4.3 Virtual Observation Points

The Virtual Observation Points algorithm is a local SAR supervision method developed from Eichfelder in 2011 [166]. The input of the VOP averaging is the S-matrices, or SAR matrix, which is a channel specific SAR estimation for each voxel in the model. The algorithm groups the calculated SAR maps into considerably fewer clusters of representative points (VOPs) that are monitored during a scan, see figure 5.15. What is crucial for the reliable performance of the VOPs is of course an accurate simulation matching closely the parameters of the subject scan.

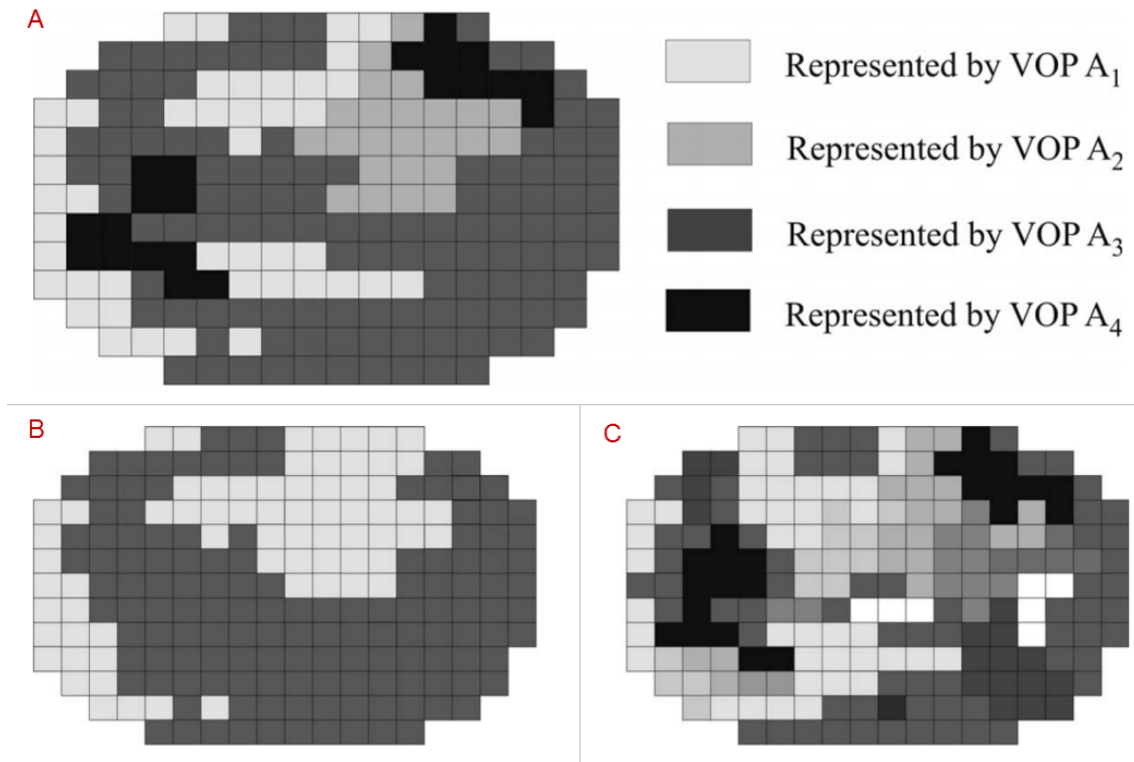


Figure 5.15: In A, the matrix is grouped in Virtual Observation Points, VOPs. Each of the few groups created share same ‘SAR’ characteristics and thus they exhibit similar local SAR under a random RF shimming combination. B) A high overestimation factor leads to two VOPs whereas in C), a smaller overestimation factor generates 5 VOPs. Reproduced with permission from [166].

Significant points to be considered about VOPs are the phase and magnitude

accuracy of first by the simulation compared to the coil and secondly the accuracy of the measurements of the directional couplers (DICO), as explained before. The possibility of significant phase errors is minimised by frequent phase calibration and system checking, virtually before every scan. As per the coil the relative phase differences as shown in table 5.2 are taken in account as discussed in the previous section.

Apart from the SAR matrix the only other input before the VOP generation is the maximum overestimation factor. This factor represents the comparison criteria elasticity between the S-matrices. Higher overestimation will make grouping more elastic and hence lead to more matrices being grouped in one cluster. Hence a higher overestimation factor leads to fewer clusters and smaller overestimation to more VOPs, figure 5.15. The table shown in figure 5.16 shows how different grids and voxel numbers compress to a few VOPs under a 5% overestimation.

Finally, data from as many phantoms as possible in order to include VOPs from smaller and bigger subjects are desirable. The dataset of the study here comprises two phantoms, an adult male and an adult female from the virtual population pack, Speag, Zurich [147, 172]. However, more representative human models are becoming available with higher resolutions which could help make the SAR monitoring via VOPs more reliable.

It is worth mentioning that the amplitude error is not more than 12% at all times and the phase variation error is less than 5° with the compression tool accounting for those errors during VOP generation [171]. In addition, the VOP algorithm would ensure that SAR will only be overestimated. The detailed mathematical analysis of the VOP algorithm is beyond the scope of this study but the reader can find more information in the original paper [166]. In the next section, the process of the SAR matrix calculation, i.e. the data input for VOPs, is provided.

		5% overestimation			
		Size of (cubical)	Number of	Virtual	Compression
		result	result voxels	Observation	
Human model		voxels (mm)	in human model	Points	factor
1	HUGO	7.18	297,897	62	4804.7
2	HUGO	4.98	891,418	68	13109.1
3	HUGO	6.72	363,682	64	5682.5
4	HUGO	6.72	363,682	232	1567.6
5	VF Duke	6.72	241,032	75	3213.8
6	VF Duke	6.72	241,032	38	6342.9
7	VF Duke	6.72	241,032	214	1126.3
8	VF Duke	4.98	586,149	65	9017.7
9	VF Duke	4.98	586,149	268	2187.1
10	VF Duke	4.98	586,149	136	4309.9
11	VF Ella	6.72	241,032	75	3213.8
12	VF Ella	6.72	241,032	152	1585.7
13	VF Ella	6.72	241,032	22	10956.0
14	VF Duke Head	4.08	1,134,246	388	2923.3

Figure 5.16: Results from clustering the SAR matrices to virtual observation points, VOPs for an 8 channel system for different anatomical phantoms and positions. The overestimation factor is set at 5% for every case. The ‘worst case SAR per unit RF excitation vector’ is in arbitrary units and hence is given for comparison between the models. The uncompressed voxel number is given along with the VOP number generated for that model along with the compression factor for each case. Adapted with permission from [166].

5.4.4 Preparation and VOP clustering

Averaging of the E-fields

For channel one, the three components of the **E**-field, E_x , E_y and E_z , are loaded with the matrices being in a+ib complex format. The averaging as described before takes place in 3 dimensions. The mesh is comprised of 1 mm sized voxels and for an average density of $1000\text{kg}/\text{m}^3$ every voxel contains 1 mg of tissue. Thus, the cubes with 3 voxels on each side correspond to about 27 mg of tissue. The averaging calculation is repeated for the **E**-fields coming from the four transmitting elements, E_1 , E_2 , E_3 and E_4 .

SAR Matrix Calculation

The definition of the S-matrices is:

$$S = \frac{\sigma_{i,j,k}}{\rho_{i,j,k}} (E_x|_{i,j,k} E_x^H|_{i,j,k} + E_y|_{i,j,k} E_y^H|_{i,j,k} + E_z|_{i,j,k} E_z^H|_{i,j,k}) \quad (5.4)$$

The above calculation is done after the averaged **E**-fields have been re-arranged in 2D matrices with one dimension being the channel number, $nC=4$, and the other the total number of voxels, $nV=761540$. In total there are three matrices, one for each field component.

$$S = \frac{\sigma}{\rho} E_x(nV, nC) \cdot E_x'(nV, nC) + \dots \\ E_y(nV, nC) \cdot E_y'(nV, nC) + E_z(nV, nC) \cdot E_z'(nV, nC) \quad (5.5)$$

The calculation generates a 3 dimensional matrix sized as $S(nV, nC, nC)$ and there are the input data for the VOP compression tool along with the overestimation factor. In this case the simulation results from a male and a female phantom were put together in those 761540 voxels that are used to calculate the SAR matrix.

VOP Compression Tool

The VOP compression tool is the last stage and is performed on the scanner computer with the input being the SAR matrix and the maximum overestimation factor. The recommended value for the overestimation factor is set to 5% [166]. The use of a small factor leads to higher calculation times on the scanner and a higher probability of aborted scans. Since it is ensured that the SAR calculation after the compression is always higher than the SAR of the full model data the selection of the factor does not increase risk.

An example of the VOP report is shown in figure 5.17. The report includes the number of VOPs against the SAR matrix voxels. In particular, figures 5.17 and 5.18 show the resultant output report from the compression tool for the simulation S-matrices at 1% and 5% overestimation factors with the second report being the one used for the local SAR supervision of the 4Tx/4Rx neck coil.

5.4.5 VOP Validation

The VOP method reduces large simulation output datasets to a few hundred monitoring points. The measured magnitude and phase from each channel are factored in to the VOP and the number calculated represents the worst SAR for the group of voxels to which the particular VOP corresponds. The highest SAR each time is the one considered for the power limitation.

The validation was performed by comparison of the worst case SAR as generated by the scanner with the worst case SAR as calculated off-line with the use of the same phase and power combination. Ten random RF shim cases were selected for the testing of the VOPs. The cases are shown in table 5.3 with the scaling factor of the magnitudes being 1.0 for each channel.

Table 5.4 shows the comparison between the VOP and the off-line calculation of the local SAR. A certain overestimation is observed for each case which is reassuring but the values seemed to be uncorrelated. The amount of overestimation can be

```

cluster 6240 contains 64 pixels (# of pixels left=124) ...
cluster 6241 contains 37 pixels (# of pixels left=87) ...
cluster 6242 contains 2 pixels (# of pixels left=85) ...
cluster 6243 contains 1 pixels (# of pixels left=84) ...
cluster 6244 contains 2 pixels (# of pixels left=82) ...
cluster 6245 contains 48 pixels (# of pixels left=34) ...
cluster 6246 contains 1 pixels (# of pixels left=33) ...
cluster 6247 contains 24 pixels (# of pixels left=9) ...
cluster 6248 contains 3 pixels (# of pixels left=6) ...
cluster 6249 contains 2 pixels (# of pixels left=4) ...
cluster 6250 contains 3 pixels (# of pixels left=1) ...
cluster 6251 contains 1 pixels (# of pixels left=0) ...
All pixels were clustered.

Checking if created UOPs are safe ...
Got 6251 UOPs and 645578 SAR matrices
.....
.....CHECK PASSED

Write UOPs to binary file: vop.dat
Coil name: Neck_4Tx4_Rx
Coil ID: Neck_4Tx_4Rx
Patient position: Head First - Supine
Total # of UOP = 6251
Maximum overestimation of local SAR (S/kg)= 0.000904445 (1 %)
'Worst case local SAR'/'Global forward power' = 4.52222

----- ALL TASKS SUCCESSFULLY COMPLETED -----

```

Figure 5.17: VOP compression tool output report for 1% overestimation factor. The SAR matrix comprises of 645578 8 by 8 2D matrices estimating the effect of each of the 8 channels of the pTx system for the particular voxel. The number of VOPs is 6251 with a global worst case estimation of 4.522 W/kg.

```

cluster 229 contains 76 pixels (# of pixels left=470) ...
cluster 230 contains 46 pixels (# of pixels left=424) ...
cluster 231 contains 26 pixels (# of pixels left=398) ...
cluster 232 contains 117 pixels (# of pixels left=281) ...
cluster 233 contains 15 pixels (# of pixels left=266) ...
cluster 234 contains 25 pixels (# of pixels left=241) ...
cluster 235 contains 91 pixels (# of pixels left=150) ...
cluster 236 contains 20 pixels (# of pixels left=130) ...
cluster 237 contains 1 pixels (# of pixels left=129) ...
cluster 238 contains 113 pixels (# of pixels left=16) ...
cluster 239 contains 2 pixels (# of pixels left=14) ...
cluster 240 contains 14 pixels (# of pixels left=0) ...
All pixels were clustered.

Checking if created UOPs are safe ...
Got 240 UOPs and 645578 SAR matrices
.....
.....CHECK PASSED

Write UOPs to binary file: vop.dat
Coil name: Neck_4Tx4_Rx
Coil ID: Neck_4Tx_4Rx
Patient position: Head First - Supine
Total # of UOP = 240
Maximum overestimation of local SAR (S/kg)= 0.00452222 (5 %)
'Worst case local SAR'/'Global forward power' = 4.52222

----- ALL TASKS SUCCESSFULLY COMPLETED -----

```

Figure 5.18: VOP compression tool output report for a 5% overestimation factor. This is the factor used for the SAR monitoring. Again 645578 SAR matrices are generated from unitary sources plus 12% due to error variation. The number of virtual observation points is now 240 and the total worst case local SAR is about 4.522 W/kg.

Degrees	Channel 0/8	Channel 1	Channel 2	Channel 3
Case 1	0°	0°	0°	0°
Case 2	0°	10°	20°	30°
Case 3	0°	20°	40°	60°
Case 4	0°	30°	60°	90°
Case 5	0°	40°	80°	120°
Case 6	0°	50°	100°	150°
Case 7	0°	60°	120°	180°
Case 8	0°	70°	140°	210°
Case 9	0°	80°	160°	240°
Case 10	0°	90°	180°	270°

Table 5.3: RF shim cases used for the VOP validation with magnitudes being 1. Each case was applied before the scan with known power per channel and the VOP 10g averaging was noted down for comparison with the off-line calculation.

Worst SAR	VOP [W/Kg]	Calculated [W/Kg]
Case 1	10.5	6.82
Case 2	10.6	7.23
Case 3	11.5	7.36
Case 4	11.9	7.25
Case 5	12.0	6.80
Case 6	11.7	6.87
Case 7	11.2	7.53
Case 8	10.5	7.07
Case 9	10.1	6.80
Case 10	10.0	7.57

Table 5.4: Scanner measurement for 10 sec VOP at the end of a 40 sec scan. The values of the VOP are contradicted with the SAR calculated off-line. There is a clear overestimation for each case from the VOP ensuring safety. However, a pattern over the two measurements could not be observed.

attributed to the error margin accounted for by the scanner that comprises the magnitude measurement error, the phase error as well as the VOP overestimation error of 5%. When combined these can indeed lead to the overestimation of 40%. The maximum overestimation was observed in case 5 where the VOP was 76% higher than the off-line SAR. The VOPs thus could be regarded a safe supervision method but nevertheless, global and worst case SAR were also calculated to fully characterise the coil.

5.4.6 Global SAR

Once local SAR monitoring has been addressed the next SAR criterion to be regulated is the global or partial body SAR defined in section 5.2.1. The gSAR is the total transmitted power over the exposed patient mass and is calculated as shown below.

$$gSAR = \frac{P_1 + P_2 + \dots + P_n [Watt]}{Exposed Patient mass [Kg]} \quad (5.6)$$

where n equals the number of transmitting elements, i.e. 4 for the examined coil.

Both global and local SAR need to be kept below the stated limits at all times and if either reaches the limit the scan must be interrupted. The SAR limits for a 6 minute scan are dependent on the ratio ‘exposed patient mass’/‘patient mass’ and for first level controlled mode the limit for any part outside the head region is 10 W/kg - (6 W/kg · ‘exposed patient mass’/‘patient mass’). Thus, the bigger the transmit array the larger the exposed mass and hence the lower will be the maximum limit. However, a separate head SAR limit has been introduced that is 3.2 W/kg for all conditions. Because the coil is a neck coil, a significant part of the lower head is within the exposed mass and for added safety the head limit of 3.2W/kg will be used.

A suitable phantom was modelled and fitted around the coil plastic cradle in order to measure the effective volume of the coil, figure 5.19. The phantom was set to be standard tissue simulant at radio frequencies with 0.7 S/m conductivity and relative permittivity of 48. The grid resolution was set at 1 mm isotropic voxels and all transmit elements were set to transmit with the same phase and amplitude. Since the system will be operated with the same power at each coil the calculation should be accurate.

The calculation of the effective volume can be performed via the use of the energy density given in joules per cubic metre. The steps of the calculation are given in the

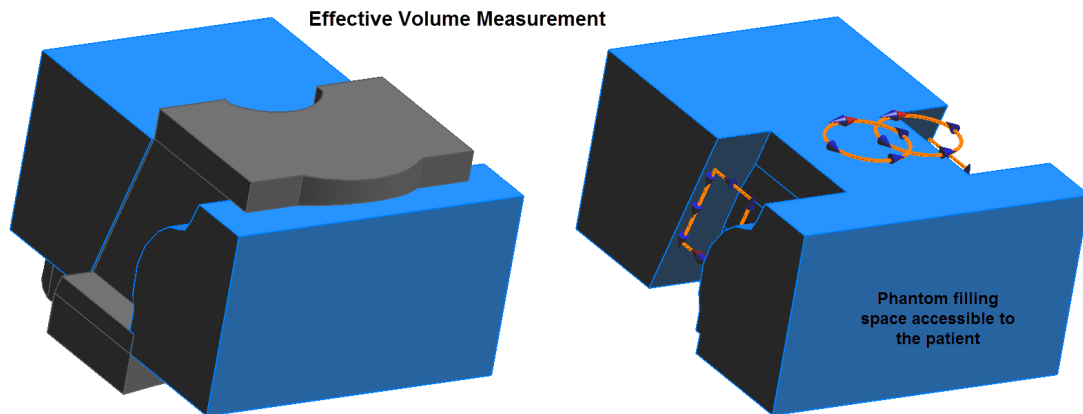


Figure 5.19: Effective volume measurement as defined in IEC60601 [167]. A suitable tissue simulant phantom was modelled in order to fit the space that would normally be covered by the patient with an additional coverage margin. A grid of 3mm resolution was used and all elements were set to transmit with the same phase and amplitude.

bullet points below. The energy density data matrix is normalised for 1 W, although this does not affect the final result.

- The energy density matrix is transformed to a 1 dimensional string array.
- The numbers in the array are sorted from the highest to the lowest values (descending order). This is done because the highest values are the ones that are located closer to the coil and the lowest ones are further away. Hence, the volume will be constructed from the closest points to the furthest ones.
- The total sum is calculated based on isotropic 1 mm voxels and the energy density given. In this particular case the total energy deposited in the field sensor is 9.5 nJ (for one wavelength) but the accuracy of the measurement is not crucial since it will not affect the volume shape.
- The voxel that sums 95% of the total energy starting from voxel one (highest to the lowest) is found via loop calculations and the corresponding energy density value is used as a mask generation criteria in the next step.
- A mask is created consisting of the values in the energy density that are higher

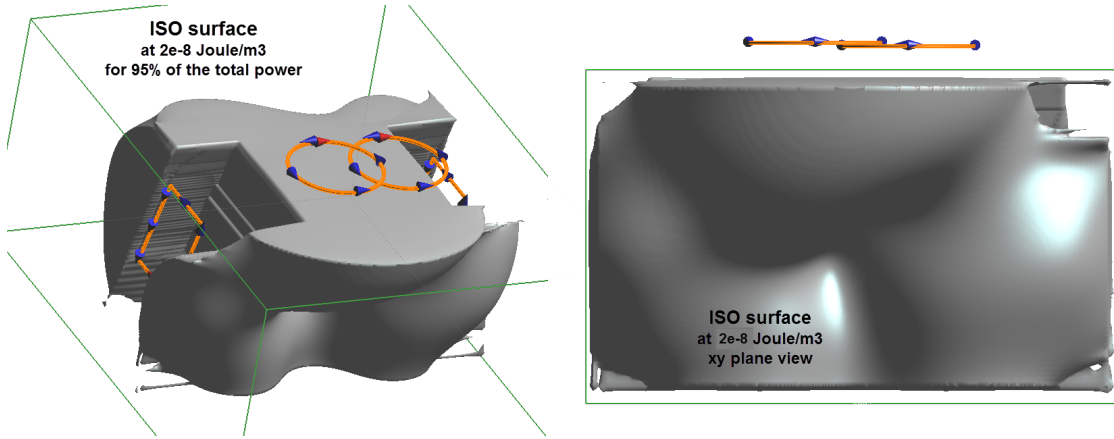


Figure 5.20: The iso margin based on the process described above was selected at $20nJ/m^3$. The volume generated is the effective coil volume and corresponds to the 95% of the total transmitted RF power in the phantom covering the space available to the patient. On the right the transverse plane is shown.

than the calculated value. The volume that is generated corresponds to the effective volume of the coil.

- A cross section of this volume with an averagely small human phantom will give the exposed patient volume.
- The volume is considered to comprise mainly of water, density $\rho = 1000kg/m^3$ from where the exposed patient mass can be extracted and used in the equation (5.6).

The matrix dimensions of the energy density were set to be 325 by 203 by 462 with 30,480,450 voxels in total. Figure 5.20 represents the volume that has energy density higher than the energy density that is found to reach 95% of the total energy dissipated per unit time.

The cross section is calculated by a multiplication of the logic mask of the Phantom with the logic mask of the iso surface for the same grid and resolution. The number of voxels that comprise the cross section of the Ella phantom intersecting with the volume that corresponds to 95% of the power is 3,825,442. That represents a volume of $TotalVolume = No_{vox} \cdot 1mm^3 = 0.003825m^3$ that for a density of 1000 W/kg given the exposed patient mass to be **3.82 kg**. An illustration of the phan-

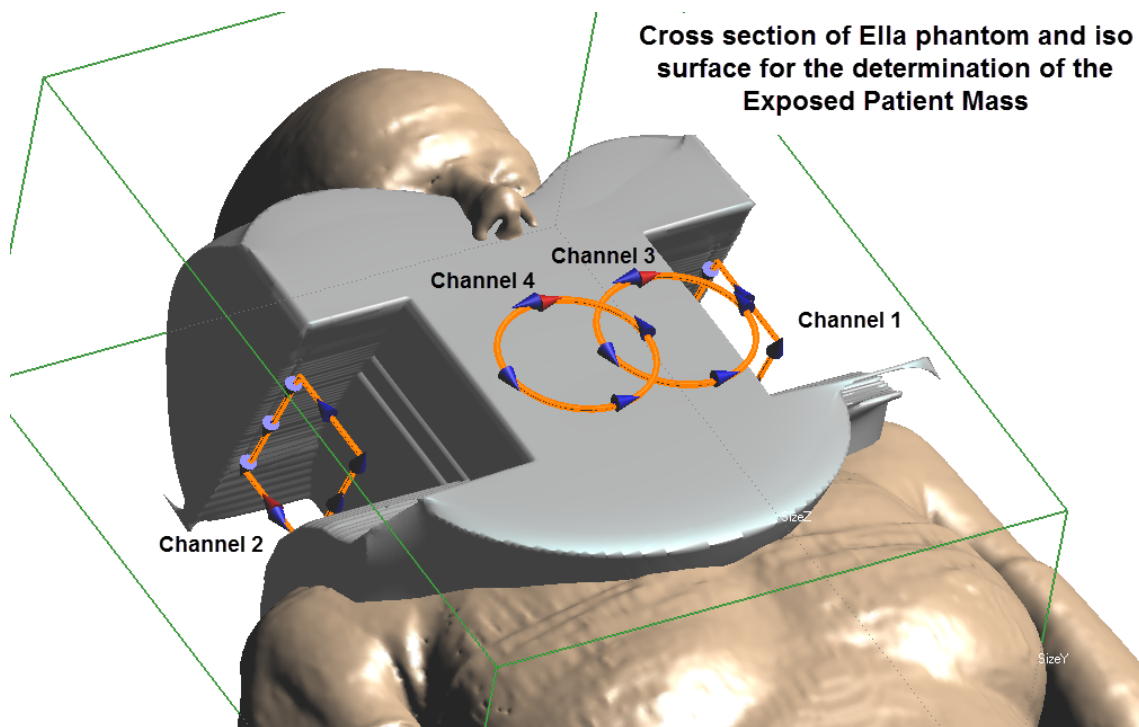


Figure 5.21: The Ella model is superimposed on the effective volume. The exposed patient mass is the mass of the phantom that corresponds to the cross section of the effective volume and the phantom volume. The resulting calculation is shown in figure 5.22.

tom, effective volume and exposed patient volume can be seen in figures 5.21 and 5.22.

Based on the calculated exposed mass the 6 minute power limit from equation (5.6) becomes:

$$SAR_g = \frac{P_{tot}}{3.82} \Rightarrow P_{tot} \leq 3.2 \cdot 3.82 = 12.22W \quad (5.7)$$

Note that the equation above does not take into account any TALES error or cable attenuation.

For a 10 second period the stated limits can be twice the 6 min power limit. However, due to the 12% power measurement variation error from the transmit antenna level sensor (TALES) system the limits are suppressed by a factor of 0.9. The cable attenuation arising from TALES to the coil plugs is set at 1.9 dB by the manufacturer and another conservative 0.5 dB (in reality it varies from 1 to 1.35 dB) from the coil plugs to the coil loops. The limit is re-adjusted based on equation (5.8).

$$P_{global} = Er_{12\%} \cdot SAR_{gl} \cdot Mass \cdot 10^{\frac{A}{10}} = 0.88 \cdot 3.2 \cdot 3.82 \cdot 10^{\frac{1.9+0.5}{10}} = 18.6 W \quad (5.8)$$

The 10 sec limit is set to be exactly double the 6 min one, i.e. about 38 W. The system will re-calculate the global SAR limit by taking into account the sliding window averaging of the TALES system, see section 5.3. It is possible to use limits per channel for 10 sec and 6 min where the consideration of the sliding window must be done by the coil developer based on the k -factor. The latter is the inverse of the exposed patient mass and denotes the global SAR imposed per unit power. The k -factor of the 4Tx/4Rx is $1/3.82$, i.e. $0.2618kg^{-1}$. The k -factor is not an accurate SAR profile characterisation for multi transmit coils, although for the global SAR and the PALI MC the calculation is adequate since it entails a total transmitted power over a specific mass. As discussed in the previous section, when local SAR is

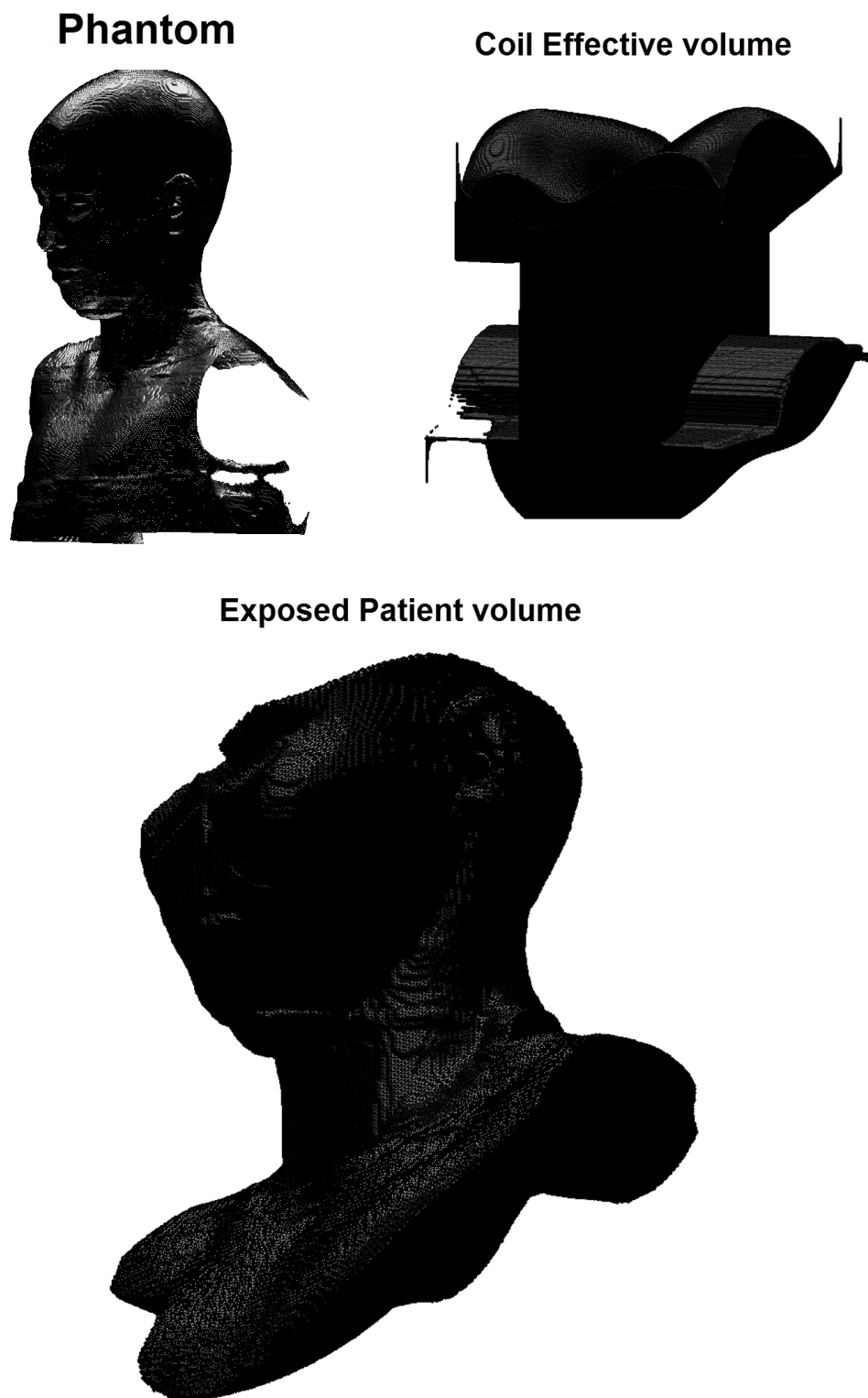


Figure 5.22: A female phantom (Ella, virtual population) was used for the calculation of the exposed volume considering that it is a smaller phantom than the male one. On the top right the iso surface is extracted as previously for up to a 95% energy accumulation compared to the total energy deposited in the material covering the patient space. On the bottom, the extracted surface is shown, which can intuitively be verified. The number of 1mm voxels that correspond to 1mg of water can give a conservative estimate of the exposed patient mass and is found to be 3.8 kg.

Global SAR	Power Limits
10 Second	37.2 Watt
6 Minute	18.6 Watt

Table 5.5: Global SAR power limits for 10 second and 6 minute periods as they have to be set by the operator for the 4Tx/4Rx neck coil.

monitored through the VOPs the global power limits can be set on the basis of the k -factor. The limits are thus summarised in table 5.5.

5.4.7 Worst Case SAR

An additional method to supervise the SAR as explained earlier is to use worst case local SAR and set power limits that would never allow a transmission scheme to reach the SAR limits. The system provides control over each TALES after the RFPA and power limits per channel can be set.

The principle behind the worst case approximation is the superposition of the **E**-field magnitudes from each channel. That way the phase element is excluded and for each voxel the highest possible **E**-field is used to calculate a worst case SAR map. The highest value from this matrix is the absolute worst case local SAR.

$$SAR_{worst\ case} = \frac{(\sum_1^n |E_i|)^2 \sigma}{2\rho} \quad (5.9)$$

The two phantoms, Duke and Ella, for 1 W power normalisation coming from each channel give a worst case local SAR of 7.0 and 7.3 W/Kg respectively. For the 6 min limits of 20 W/Kg the allowed power factor of a continuous wave would be $20/7.3 = 2.74$. Since the calculation is done for 1 W per channel the factor is applied to 1 W and the total allowed power per channel is 2.74 Watt. The power allowed per channel for the 10 sec limits is double the value, i.e. 5.48 Watt. Table 5.6 summarises the limits accounting for 10% error variation in TALES measurement.

The worst case local SAR can be a conservative method of ensuring RF safety and in addition more straight forward. Since the power limit per channel is 2.74 W

Local SAR	Power Limits per channel
10 Second	4.93 Watt
6 Minute	2.47 Watt

Table 5.6: Local SAR power limits for 10 second and 6 minute periods as they have to be set by the operator for the 4Tx/4Rx neck coil. Both global and local power limits are activated for any scan.

the maximum total could be no more than 10.96 W. Since the global power was set before at 18.6 W is higher than the ‘per channel’ the latter will only supervise the SAR.

5.4.8 Experimental Validation of the Coil

The coil was validated with temperature measurements in a phantom with known power output and duration. The temperature increase rate is an indication of the power deposition rate as long as the specific capacity, c_m , of the sample is known. The particulars of the phantom selection and the simulation thermal solver used are discussed in the next section.

A 1.9 kg piece of lean beef meat was selected to be the phantom of the simulation and could load the coil sufficiently. A temperature probe was inserted 10 mm deep in the beef below the top coils. The total distance between the top coil plane and probe was 50 mm. A series of scans with different power outputs were run and compared with the simulation for validation of the coil. See figure 5.23 for the experimental setup and the model constructed in Semcad-X environment.

Top Channel Hot Spot

A scan was run for each element individually for 10 and 20 W continuous wave and another 4 scans with incremental power for all channels transmitting with the same phase and magnitude. The power was controlled by adjusting the voltage of the RF pulses on a high duty cycle sequence and was confirmed by the combined TALEs readings (PALI MC). The sensor positioning in the simulation was selected in order

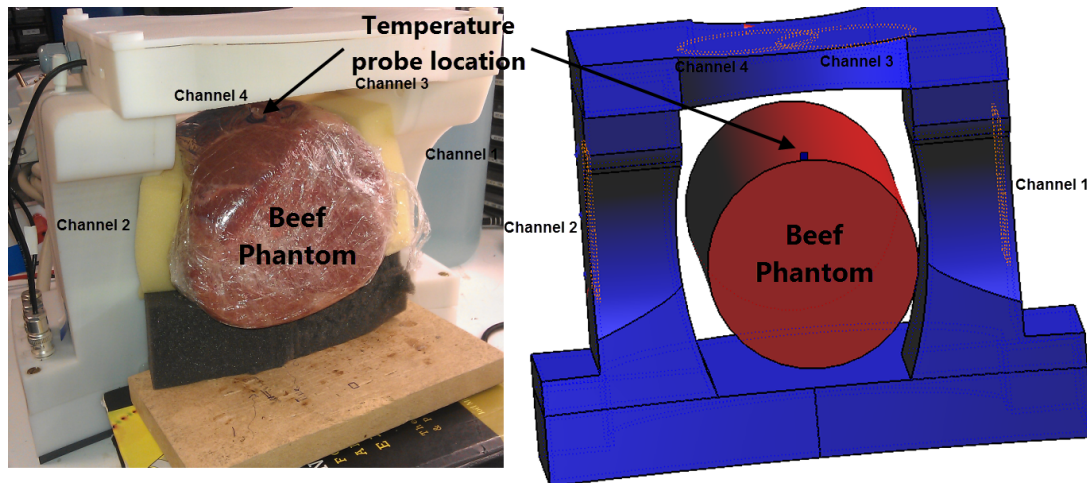


Figure 5.23: Photo of the experiment setup (left) and the corresponding Semcad-X constructed model. The diameter and length of the phantom were selected to match the particular geometry and weight of the purchased meat. The temperature probe was positioned roughly 10 mm deep with a thin cut and was placed under the overlap of the loop 3 and 4.

to match with the heating rate from the measurements. The sensor is regarded reliable only if it can reproduce the measured values for all four channels. Figure 5.24 shows the process followed to validate the sensor location. The experiment and simulation results for each case are summarised in table 5.7. The complete temperature data is given in figure 5.25.

Side Channel Hot Spot

Another experiment was performed with the same phantom shape, composition and weight in order to assess the heating only from the side coils. The sensors were placed on the expected hot spot for the right and left channels. Table 5.8 summarises the results of the simulated heating versus the measured temperature. The absolute increase is comparable with a deviation that can be attributed to a reasonable error accounting for the 0.1°C resolution of the temperature kit. Figure 5.26 shows how the heating looks in simulation and the maximum temperature measured in the hot spot.

Before the experiment an effort was done to locate more accurately the hot spot of

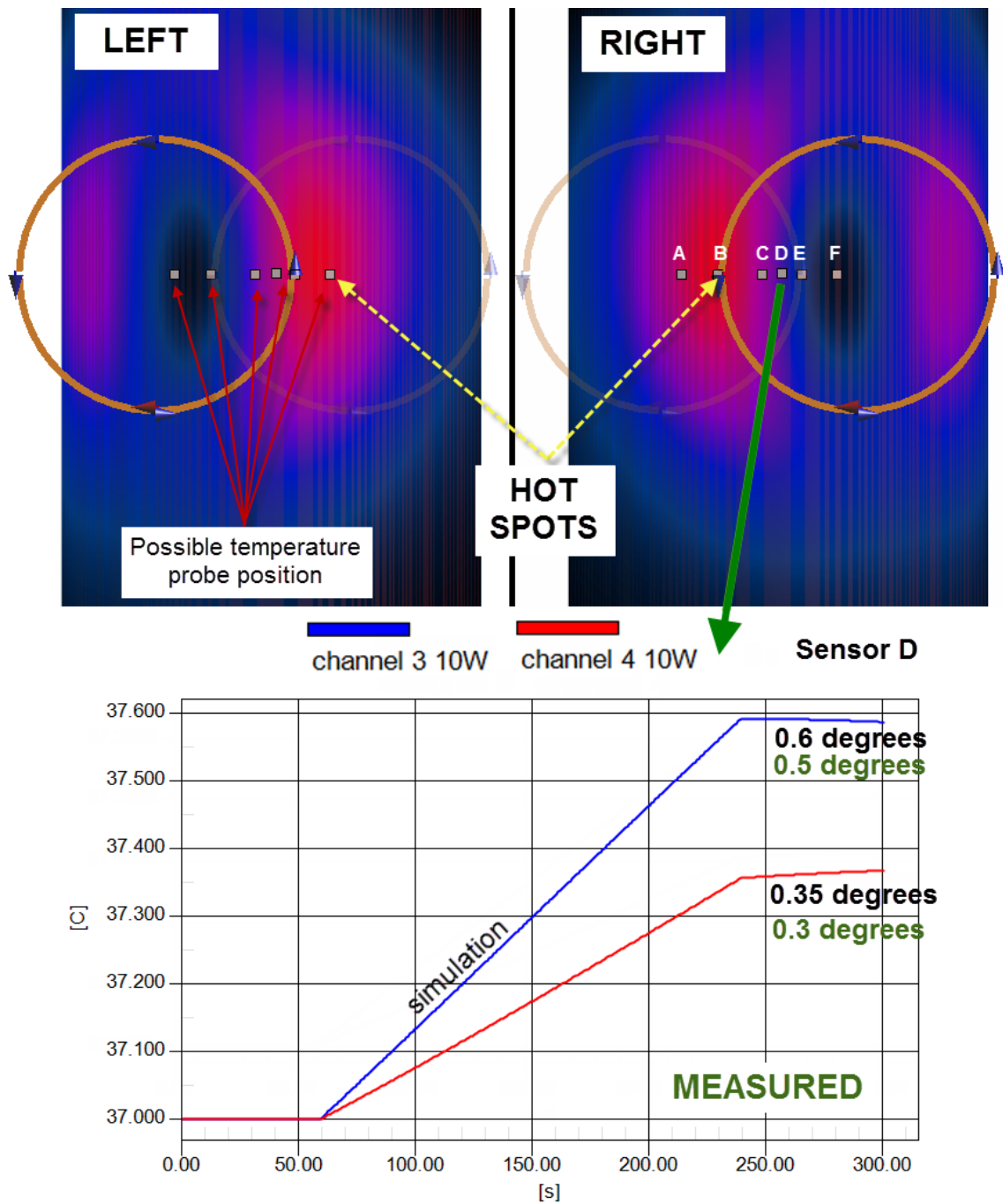


Figure 5.24: The upper figure shows the coronal surface heating profile of the phantom under the top loops. Inevitably, the variability of the real experiment compared to the simulation can lead to discrepancies in the measured temperature. The possible sensor locations are shown above with the sensor D being the most probable location used during the experiment. That is because the ratio between the heating from the two channels matches for any power selected.

Simulation vs Measurement for 3 min scan			
10 Watt	Thermal solver [$^{\circ}C$]	Measurement [$^{\circ}C$]	Overestimation [$^{\circ}C$]
channel 1 (1)	0.10	0.2	-0.10
channel 2 (2)	0.10	0.1	0
channel 3 (3)	0.60	0.5	0.10
channel 4 (4)	0.35	0.3	0.05
20 Watt	Thermal solver [$^{\circ}C$]	Measurement [$^{\circ}C$]	Overestimation
channel 1 (5)	0.15	0*	0.15
channel 2 (6)	0.15	0.10	0.05
channel 3 (7)	1.20	0.90	0.30
channel 4 (8)	0.70	0.40	0.30
All channels **	Thermal solver [$^{\circ}C$]	Measurement [$^{\circ}C$]	Overestimation
4X5W (9)	0.35	0	0.35
4X10W (10)	0.70	0.4	0.30
4X15W (11)	1.04	0.5	0.54
4X20W (12)	1.39	0.6	0.79

Table 5.7: Summary of results for the temperature measurements on the surface of a beef phantom positioned as shown in figures 5.23 and 5.24. The number in parentheses denotes the scan as shown in the temperature data, figure 5.25. The values agree in most cases with the desired overestimation from the simulation adding safety margin to the worst case scenario. Overestimation was observed in each case which is reassuring as far as the safety is concerned.

* For 20 W transmission the channels 1 and 2 lost their matching due to PIN diodes being subjected to power beyond their parameters.

** Matching of channels 1 & 2 is lost due to excessive power in the 20 W experiment. Channels 3 & 4 are only contributing in the heating of the phantom.

the side coils and hence a heating profile test was done with the use of a temperature sensitive liquid crystal sheet. The transmission was 25 min for 10 W CW with no significant change in the colouration of the heating surface. The relatively low power used for the component protection was not adequate to generate a thermal increase within the sensitivity of the film used.

Surface Profiles

The simulated thermal profiles from each channel are shown in figure 5.27. A side view of the side channels is shown in figure 5.28. However, the crystal sheet for the side channels could not generate any observable pattern as described and shown in

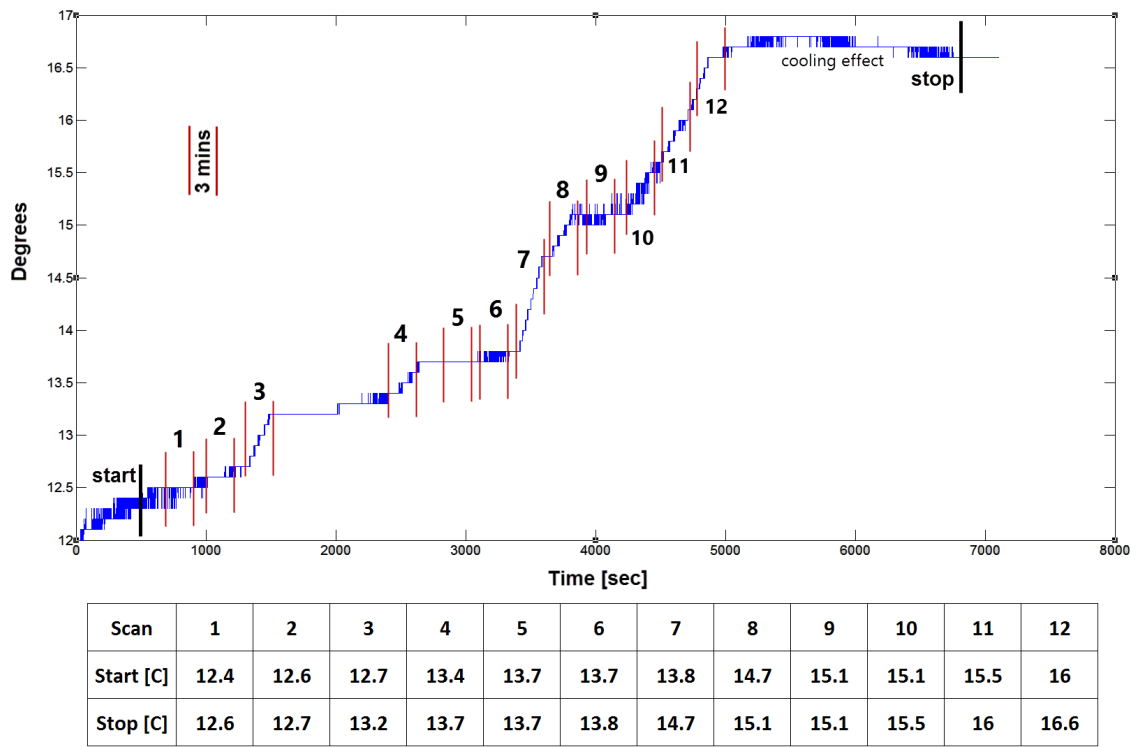


Figure 5.25: The number corresponds to the experiment number as marked in the table 5.7. Each scan lasts 3 minutes, between the two red vertical lines. Some time was allowed between scans for the settling of the thermal profile. The heating profile was taken after the last scan.

figure 5.29. The heating profile for the top coils is shown in 5.30 and the heating rate is higher than that of the side coils. The receive array position in front of the side transmit loops is partially blocking the $\mathbf{E}$ -field even though it is actively detuned. The $\mathbf{B}_1$ field is also affected but the SAR reduction is higher in proportion and thus the configuration is deemed as a safer design, i.e. higher $\mathbf{B}_1$ -to-SAR ratio. The concept will be discussed in the next chapter.

In figure 5.30 the transmission was performed with all channels at a maximum power of 20 W each, i.e. 80 W total power for a duration of 10 minutes. The sample was taken out of the magnet bore for photography and the specifications of the liquid crystal sheet can be found in subsection 5.6.1.

Simulation vs Measurement for 3 min scan			
10 Watt	Thermal solver [$^{\circ}C$]	Measurement [$^{\circ}C$]	Overestimation [$^{\circ}C$]
channel 1	0.33	0.15	0.18
channel 2	0.31	0.15	0.16

Table 5.8: Heating results for the right and left channels with the temperature sensor placed on the expected hot spot. The power transmission was at 10 W CW for component safety. The measurement for each case was repeated twice and the average compared with the simulated temperature increase.

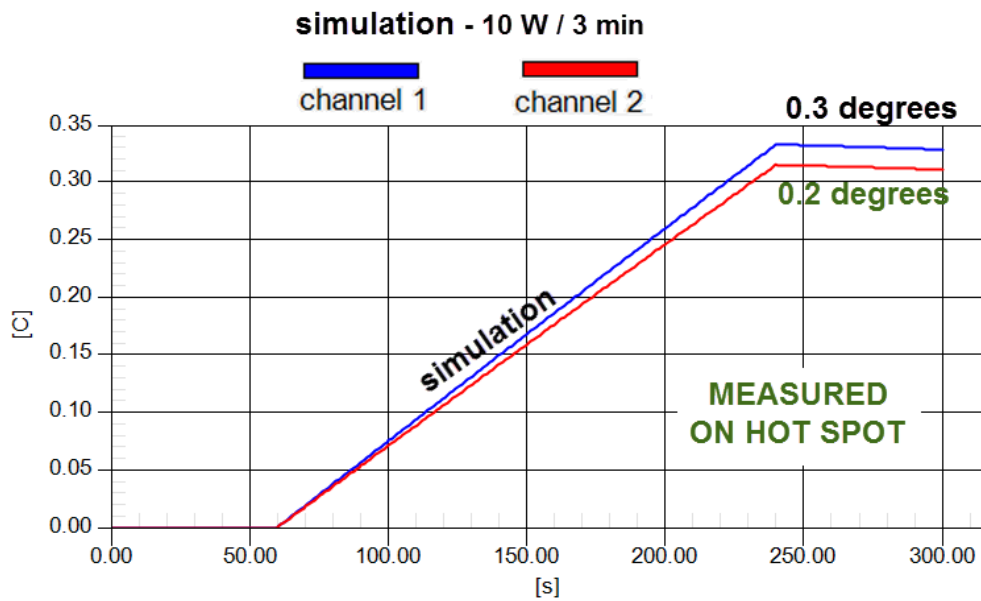


Figure 5.26: The left and right channels (1 & 2) were set to transmit, in two separate scans, at 10 W continuous wave for 3 minutes. The expected temperature increase was 0.3 degrees for both channels and the measured on the approximate hot spot location was 0.2 degrees. See table 5.8.

5.5 Other Safety Concerns

5.5.1 Common Mode Currents

Common mode currents running via the outer shield can be a hazard for a patient who comes into skin contact with cables during a scan. The currents will eventually deposit heat in the resistance of the shield and start to raise the temperature of the cable. The best way to address this is to use a cable trap which is a high impedance in series with the shield. The particulars of this solution have been discussed in the previous chapter and a practical implementation used here can be seen in figure

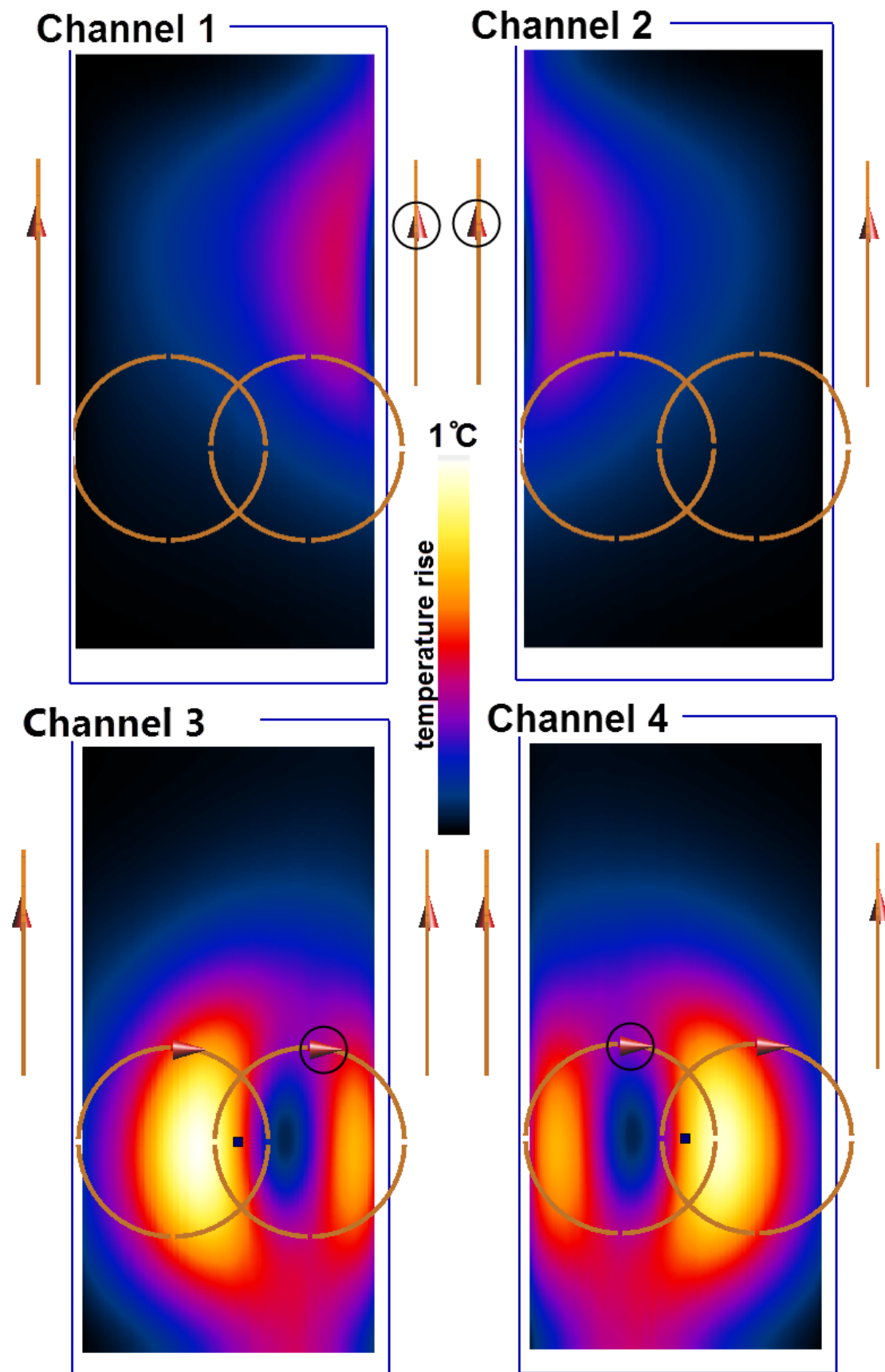


Figure 5.27: Heating profile generated by the thermal solver on a phantom with similar properties to beef. The thermal profiles of loops 3 and 4 can explain the discrepancies observed between simulation and measurement for loop 3. For a small deviation in location a fairly large variation in thermal heating is observed.

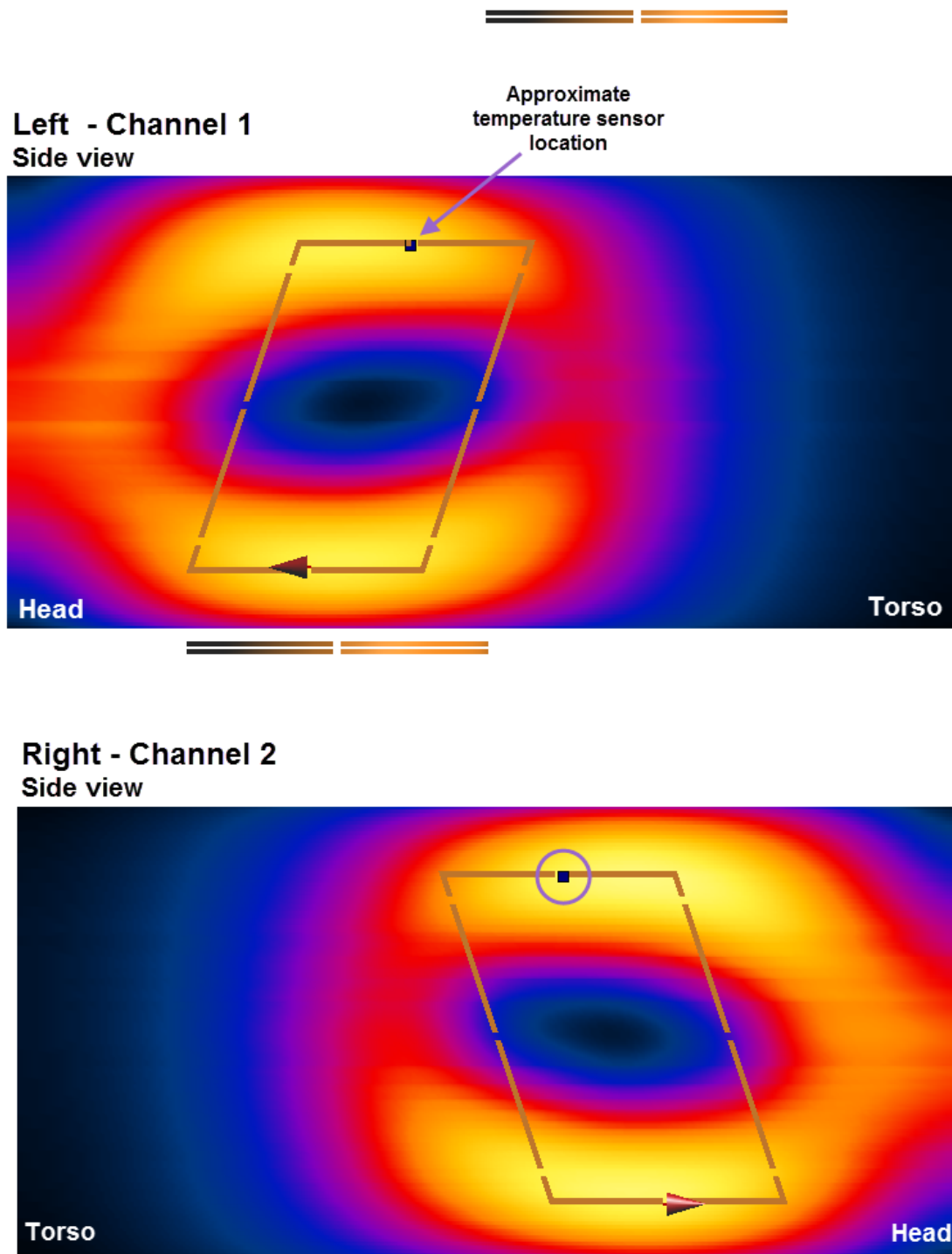


Figure 5.28: Side view of the estimated surface heating profile from the side transmit channels 1 (top) and 2 (bottom). The blue point indicates the sensor location used in about 10 mm depth in the computational phantom, and the approximate position of the optic fibre probe sensor during the experiment with the meat.

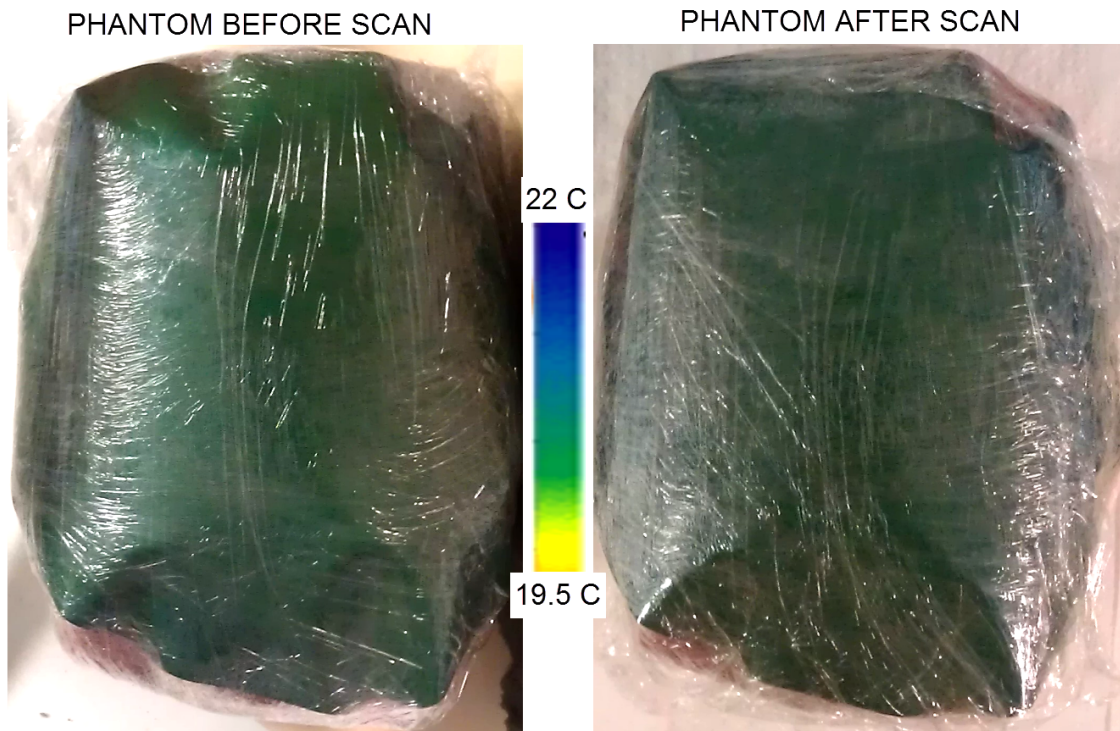


Figure 5.29: Before the experiment with the side channels a heating profile test was performed. A separate beef phantom was left to reach approximately 20° , hence a green hue, and was positioned directly on the side coil without any padding. After 25 minutes of 10 W CW no significant temperature increase was observed with the sensitive film keeping its initial colour.

5.31.

An effective way to investigate whether the outer shield of the cable is heating is to use temperature sensitive stickers. Different temperature sensitive areas will turn black in the event that the stated temperature is exceeded. During the scans described above multiple stickers were placed at the cable and close to the coil where there might be a probability that the cable could come into contact with the skin. By the end of the test scans there were no black areas on the sticker and hence the temperature of the cables stayed for the whole experiment below 37°C and were therefore safe for skin contact, figure 5.32.

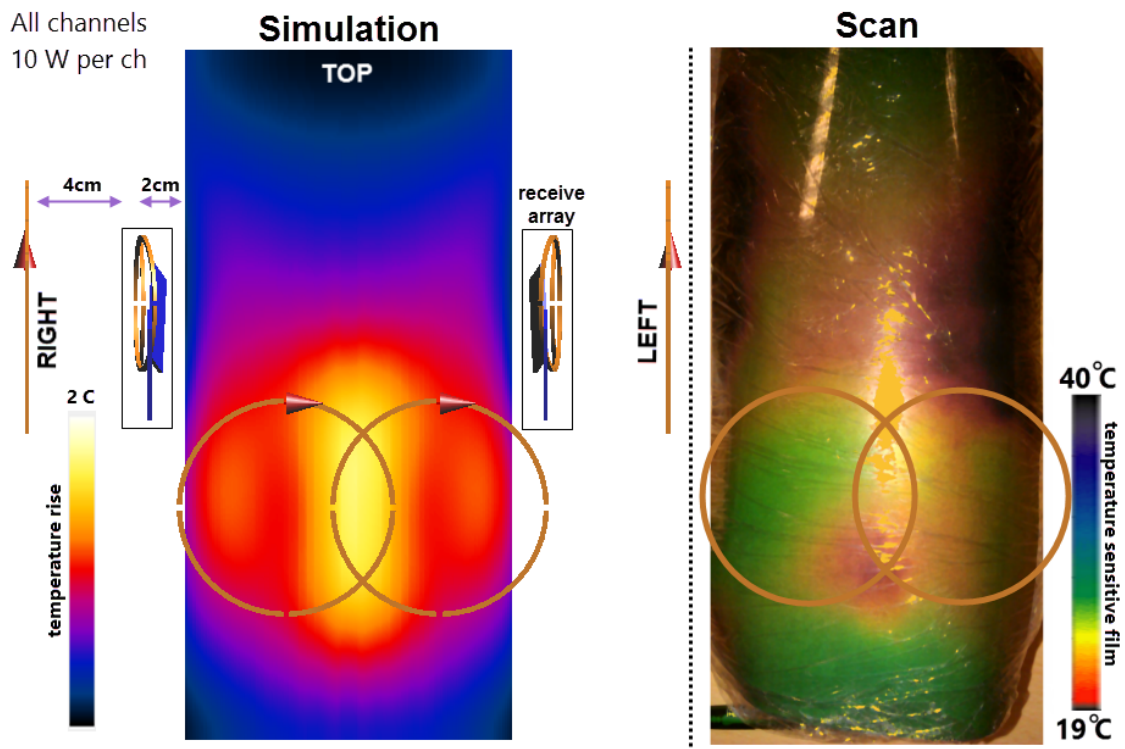


Figure 5.30: Heating profile after a 10-minute scan with 10 W continuous wave power from each channel, i.e. 40 W at the PALI MC. The profiles approximate each other reasonably. Most of the heating is deposited by the two top coils due to the **E**-field reduction observed by the receive array.

5.5.2 Exposed Metallic Parts

Any coil needs to be safe for the patient and ideally have no exposed metallic parts such as variable capacitors for matching and tuning. On the 4Tx/4Rx coil the only exposed metallic parts are the BNC connections of the top coil to the main cradle. These are electrically grounded for RF and DC and are deemed safe for the patient during scanning. Also, holes to reach variable capacitors were avoided at any part in the coil by employing plastic tuning rods.

5.6 Experimental Validation of Simulations

The simulation verification is an important step in RF safety and thus validation of the results takes place in the scanner using temperature measurements. Phantoms with known properties were used and the local temperature increase was compared

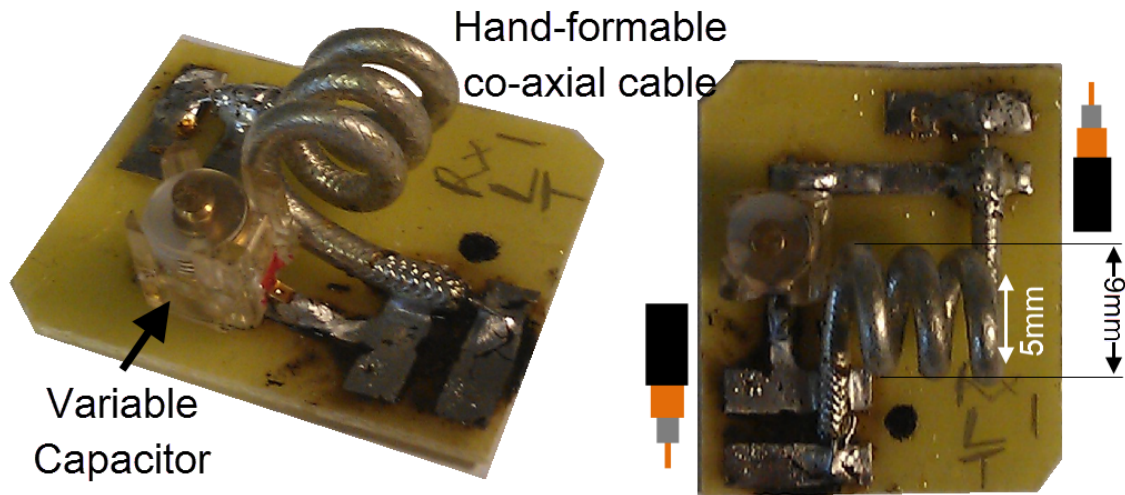


Figure 5.31: The shield trap used for the coil construction is shown here. This circuitry can be placed between any co-axial cable in order to suppress shield currents at the selected frequency. This way, coupling between the shield and the environment is minimised and cable heating is avoided.

to the simulation. A single and a dual loop coil along with a meat phantom were used for the experimental measurements. Once the results have been validated the simulations are considered accurate and reliable coil safety analysis. The experimental validation of the coil performed before did not examine the validity of the simulation but only the validity of the coil construction.

5.6.1 Methods

Two copper loops were designed from each side with their tuning and matching capacitors, see figure 5.33. The load was selected to be a lump of beef meat in the simulation and the experiment, figure 5.35. The model phantom was designed so that it was geometrically similar to the one purchased from the local store, figure 5.39 on the left.

A 297.2 MHz broadband simulation was set up and run multiple times for every element until the coils were sufficiently tuned and matched to less than -15 dB. The **E**-fields were extracted with normalisation of 1 W, i.e. disregarding the possible mismatch, as discussed before. Following the simulations, the actual coils were

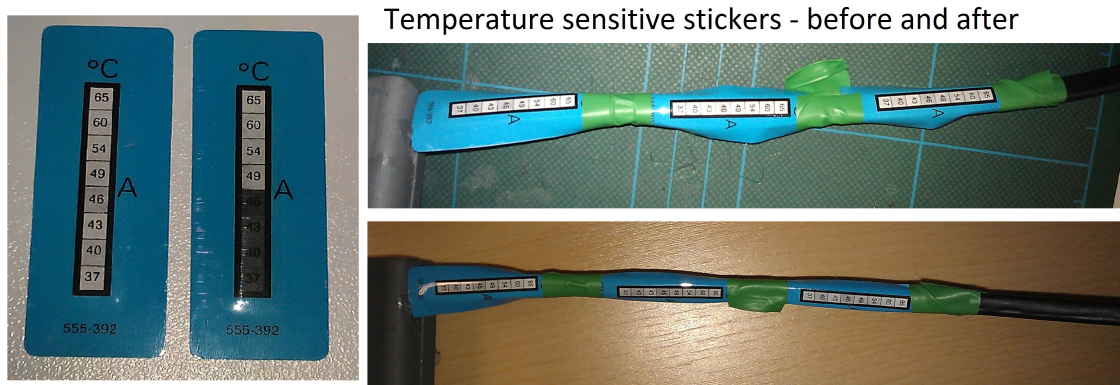


Figure 5.32: Temperature sensitive stickers on the left as an example of the effect that takes place when the stated temperature is exceeded. The right panel shows the cable before (uppermost) and after (lowermost) the test MRI sequences. Note that there was no blackening observed.

tuned and matched to less than -20 dB before the scan.

A section of beef leg joint was selected as a phantom and for the measurement of the temperature since it has thermal and electrical properties very similar to human muscle tissue, table 5.9. The phantom type would not affect the validation result since it is a comparative process, but it is useful to use a material close in properties to a human tissue in order to observe its behaviour in strong RF fields. Beef meat was also preferable since it has convenient thermal properties that are a high specific heat capacity and low thermal conductivity compared to water. These two properties translate in higher temperature ‘inertia’ and slower heat diffusion within the material; the two combined can make comparisons more reliable.

Fibre optic probes were used (Qualitrol Neoptix, Quebec, Canada) and placed on the surface of the meat in a shallow cut, figure 5.35. Two sensors were used for the dual loop experiment. A temperature sensitive liquid crystal sheet was laid on the top, beneath the transmitting coil in order to visualise the heating pattern and locate hot spots. The colour of the sheet was dependent on the temperature according to the function in figure 5.34 provided by the manufacturer. The attention in the crystal sheet experiments was put on the heating pattern rather than on the determination of the surface temperatures which were deemed unreliable.

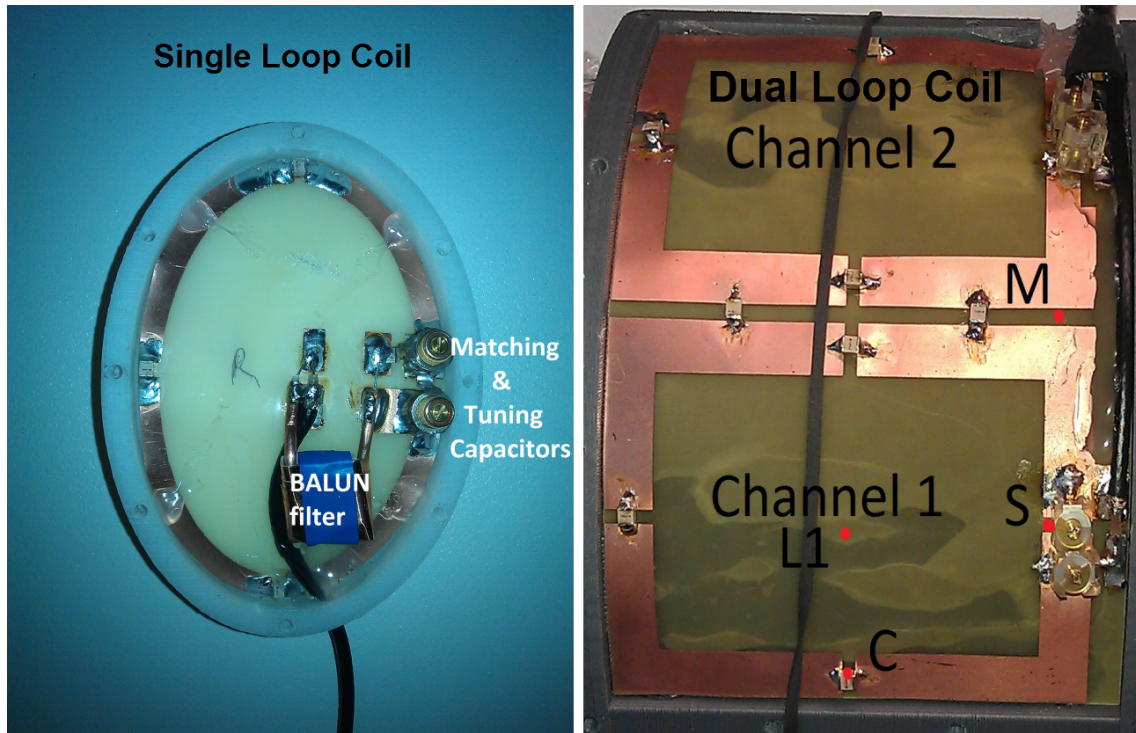


Figure 5.33: Single loop coil on the left, dual loop coil with decoupling capacitors on the right. Both coils are one side of two separate neck coils with identical loop design and cradle for the coverage of both sides of the neck. Thus, the left coil is part of a 2-channel coil and the right is part of a 4-channel neck coil.

The size of the specimen was roughly 90mm by 130mm by 150mm with an ellipsoid geometrical shape and the mass was 1.5kg. A Free Induction Decay (FID) sequence was loaded with $TR=200ms$, RF pulse width, $\tau = 2ms$ and thus 1% duty cycle with input RF voltage variable to manipulate the total power output. The continuous wave equivalent input power was $P_{in} = V_{in}^2/50 \cdot DC(1\%)$ with -0.5dB due to cable losses. A fixed 170 V rms input was set so that the power output of the coil was 5.2 W, table 5.10.

The phantom was prepared on the bench by removing the excessive fat and placing it in a food nylon wrap. The appropriate marking was made so that positioning of the coil and the temperature probes was accurate. The sensor was placed below the source and tuning capacitor of the single loop and the two sensors for the dual loop were placed below the two sources. The coil was slid sideways to change location of the temperature measurements relative to the coil and thus to provide more

	Relative permittivity (ϵ_r)	Electrical conductivity σ [S/m]	Specific Heat Capacity (c_m) [J/Kg/K]	Thermal Conductivity (k) [$W/m/K$]
Beef	58	0.77	3180	0.45
Fat	11.7	0.08	2348	0.21
Muscle	58.2	0.77	3421	0.49
Salt Water	80	4.8	4180	0.6

Table 5.9: Thermal and electric properties of beef, fat, tissue muscle and salted water. Due to the fairly similar properties of the beef and muscle the former was selected to be the phantom for the experimental measurements to observe the heating patterns.

Specific heat capacity values were taken from engineeringtoolbox.com and the rest in an approximation based on the fat and muscle values as provided by the virtual family material list. Thermal conductivities were determined with the use of the same website.

Type	FID
TR/TE	200ms/1.1ms
RF [τ]	2ms
RFin [V]	170V
Duty Cycle	1%
CW power	5.2 W (-0.5 dB loss)

Table 5.10: Sequence used for temperature measurements

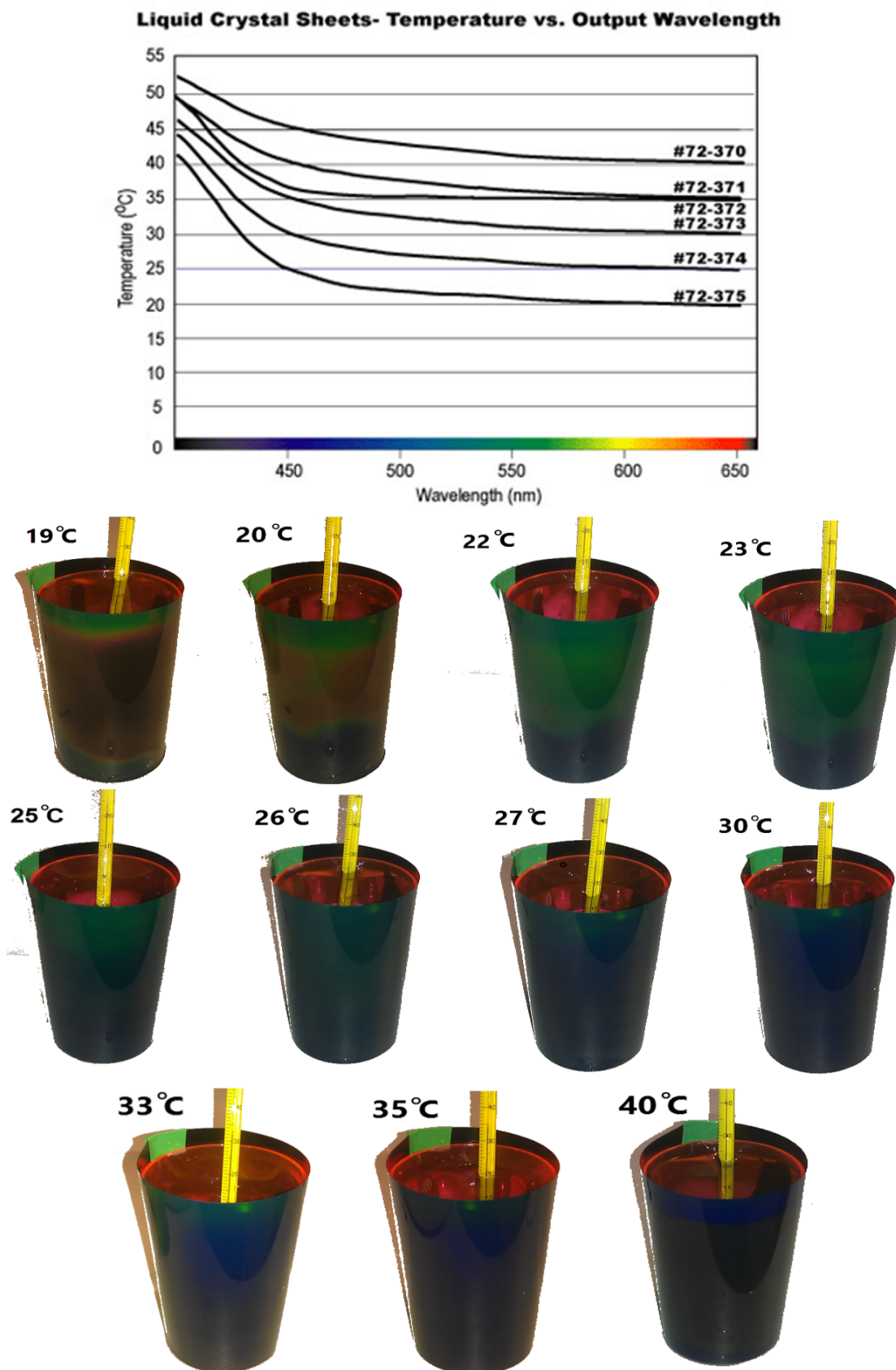


Figure 5.34: Upper panel shows the sensitivity diagram of the 72-375 liquid crystal sheet that was used in this experiment; it is particularly sensitive in the region between 20 and 25°C. Lower panel shows a piece of the liquid crystal sheet being wrapped around multiple plastic glasses filled with water with known temperature. The camera and photographic conditions are the same as the ones used after the scan.

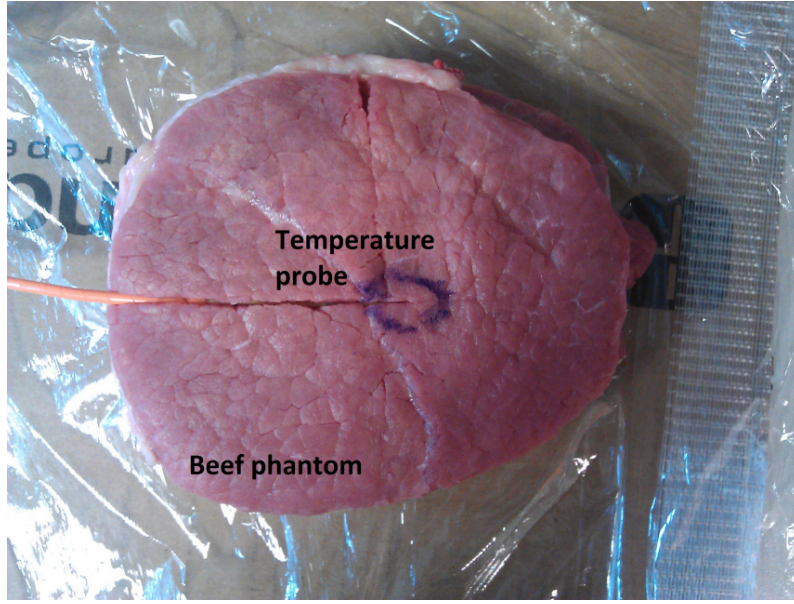


Figure 5.35: Preparation of one of the beef phantoms that were used for the experimental validation. The blue circle shows where the temperature measurement was taken by the fibre optic probe. The depth of the cut was approximately 10 mm.

data if necessary.

The two temperature sensors were placed under the two sources as strong **E**-fields are expected in that area. The effect of each of the two channels was recorded separately as well as their combination in pTx mode. For every measurement the FID sequence duration was 2 minutes with a CW equivalent power of 5.2 W. For each scan, the temperature increase was recorded. Measurements for channel 1, 2 and combined were performed twice with different initial temperatures.

After the acquisition of the temperature, the model and the setup were reconstructed in Semcad X (SPEAG, Zurich) with temperature measurement points placed in similar locations to the real ones. The thermal solver in Semcad X solves the Pennes bioheat equation (5.10) [173].

$$\rho c \frac{dT}{dt} = \nabla \cdot (k \nabla T) + \rho Q + \rho S - \rho_b c_b \rho \omega (T - T_b) \quad (5.10)$$

where ρ is density, k is the thermal conductivity, S is the specific absorption rate or SAR, ω is the perfusion rate and Q is the metabolic heat generation rate; ρ_b , c_b ,

and T_b are the density, specific heat capacity and temperature of the blood.

Each term in the Pennes equation represents a physical process of heat transfer such as convection, diffusion etc. However, on this occasion we exclude the effects of the heating mechanisms except for the 3rd term which refers to the heating from power deposition in the tissue or RF heating. Equation (5.10) can be simplified to:

$$SAR = c_m \frac{dT}{dt} \quad (5.11)$$

from which the SAR can be extracted by measuring the rate of temperature change during a measurement.

Replicating a temperature monitoring experiment in the computational environment can be challenging due to the limitations of the CAD software, the discrepancies between the actual structure and properties of the phantom and even the placement of the virtual temperature probes relative to the coil. The purpose of this experiment is to try to eliminate the previous factors by concentrating on the effect that different power levels have in the phantom. The hypothesis is that the spatial map of the normalised effect from a fixed power increase should be the same in the measurement and simulation.

With regards to the simulations, the **E**-fields were extracted and the approximate point on the phantom was examined. The points of interest were selected based on the expected maximum power deposition or ‘hot spot’, figure 5.36. These points were chosen to be under the sources, on the surface of the phantom. Finally, based on the thermal properties and the **E**-fields (normalized to a target power), the thermal solver provided temperature change estimates.

Cooling Effect

As expected, the temperature increase is not linear for the duration of the transmission but follows an exponential profile as seen in figure 5.37, where the heating rate decreases as the temperature of the phantom becomes higher. In the same fig-

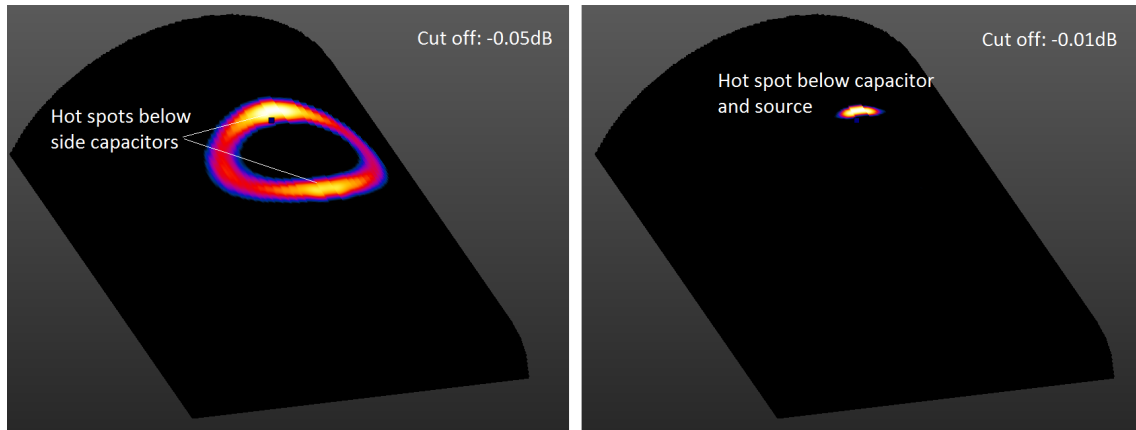


Figure 5.36: Images of the surface heat distribution as estimated by the thermal solver, with a cut off temperature in order to determine the global pattern (left) and the hot spot (right). Left images shows that the heat is deposited in a pattern similar to the loop. The right images shows that the capacitors are the areas of increased temperature due to the $\mathbf{E}$ -field presence and in particular, the hottest spot is below the source and tuning capacitor.

ure, the absolute temperature increase in 10-minute blocks shows that for the same power deposition the temperature increase after 40 min is only one quarter of that absorbed in the first 10 minutes, table 5.11.

The absolute RF power deposition does not change but the increasing temperature difference between the surrounding environment and the phantom results in higher levels of heat extraction. In the meat phantom there is arguably no other means of heat regulation other than the thermal diffusion but this is far from true for living organisms. To estimate the SAR, scanning durations of 2 to 4 minutes were selected since the ‘cooling effect’ has minimal influence in this early phase, making comparisons with the simulation more reliable. It is worth mentioning that the thermoregulation capacity of living organisms is very complex and recent studies have shown that the heat extraction increases up to 30-fold when the body temperature increases beyond normal levels [174]. If this is even only partially true then SAR would be calculated in a conservative ‘overestimation’ regime, which is desirable.

5.2 Watt	Initial [°C]	10 min	20 min	30 min	40 min	50 min
L channel	23.9	25.5	26.7	28.1	29.1	29.5
Increase		1.6	1.2	1.4	1	0.4

Table 5.11: Temperature increase over a constant transmission of 5.2 W for 50 min. Temperature was sampled every 10 mins. Due to cooling effect, the temperature rate decreases over time for constant power transmission.

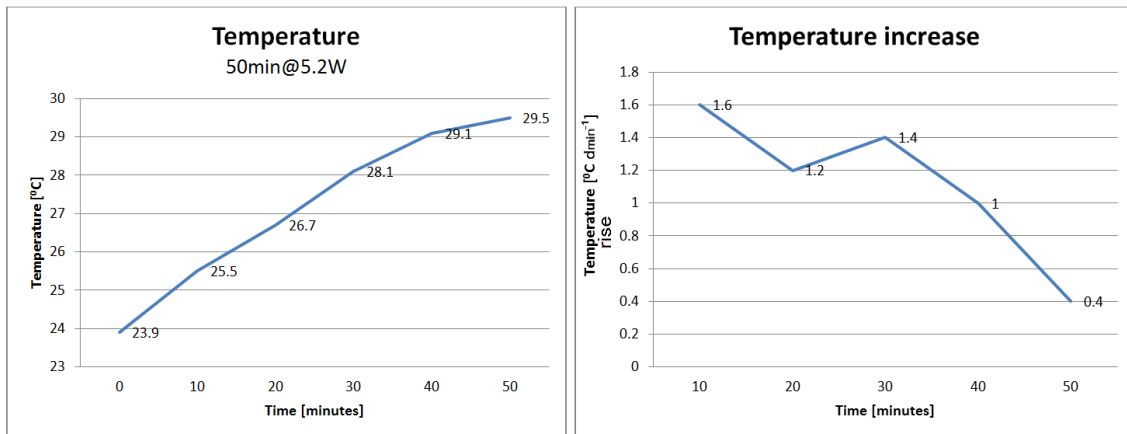


Figure 5.37: Left: Temperature increase over a 50-minute transmission period. The heating curve follows a decaying exponential pattern rather than a linear one due to cooling effect. On the right, the rate of temperature is shown for each of the 10 min intervals. For the same amount of power deposition over each interval the rate of temperature is different and decreasing as expected.

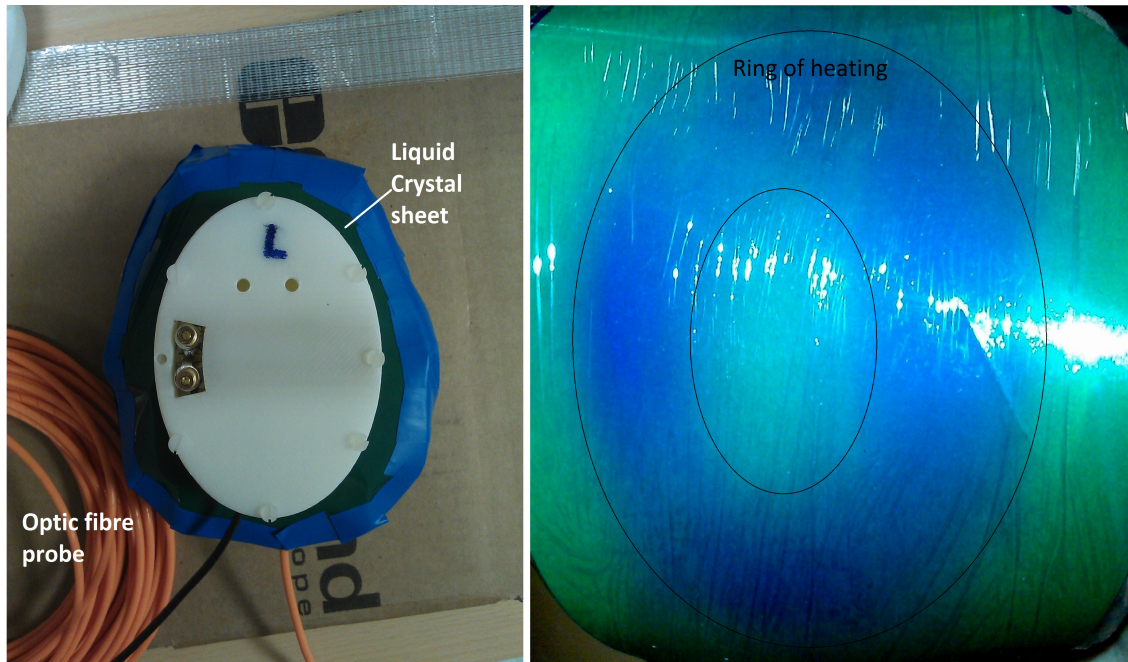


Figure 5.38: Liquid crystal sheet just after scan of 20 min @ 5.8 W transmission. The left coil was placed on the top of the sheet and the sequence was run until the temperature in the sensor lay well inside the range of the crystal sheet ($20 - 25^{\circ}\text{C}$).

5.6.2 Results

Single Loop Measurements

In the first set of results, the heat distribution for the single loop is presented, see figure 5.38 with the corresponding simulation shown in figure 5.39. One additional limitation of the use of liquid crystal sheets is the probable loss of colour contrast while the phantom is taken out of the magnet for photography. The colour scale of the crystal sheets is not linear so it is not straightforward to compare it with the simulation maps. However, it was found that there was a visible approximation both in the simulation validation experiments and the 4Tx/4Rx coil validation.

A first measurement entailed a simulation of 1 W heating over a 4 minute period for a suitable phantom. The result can be seen in figure 5.40 where the calculated increase in the hot spot is 0.25°C . In table 5.12 the measurements for two scans are given and the average increase is 0.2°C . The estimated SAR is 5 W/kg, very close to the simulation value of 6.25 W/kg. The resolution of the temperature probes is 0.1°C

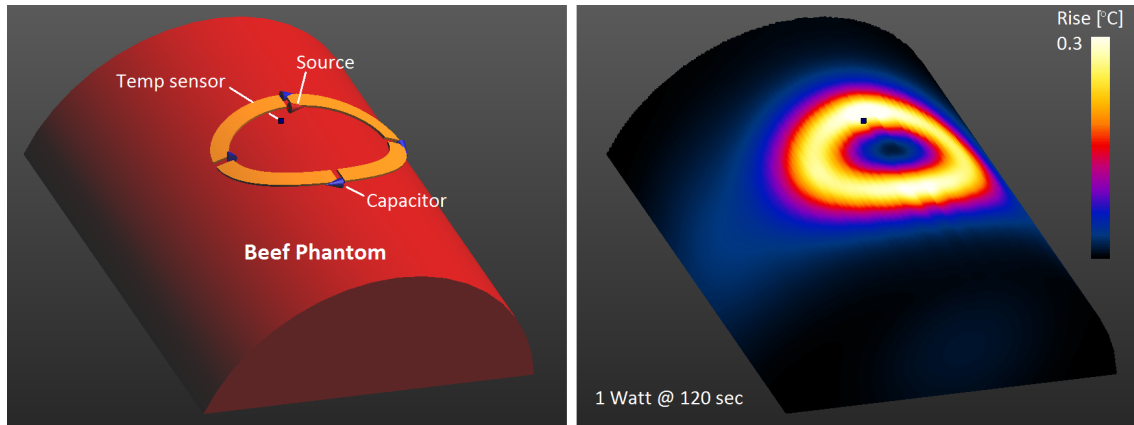


Figure 5.39: The heating profile as calculated in Semcad X. The scale is of the same range as the liquid crystal sheet but here it is linear as opposed to a slightly exponential relation in the sheet.

and the expected temperature rise for the specific power is less than 1°C . For this reason it was determined that 1 W is a relatively low power to draw conclusions for the accuracy of the simulations due to the small temperature increase with respect to the resolution of the temperature reading device. Further scans were performed with higher power input.

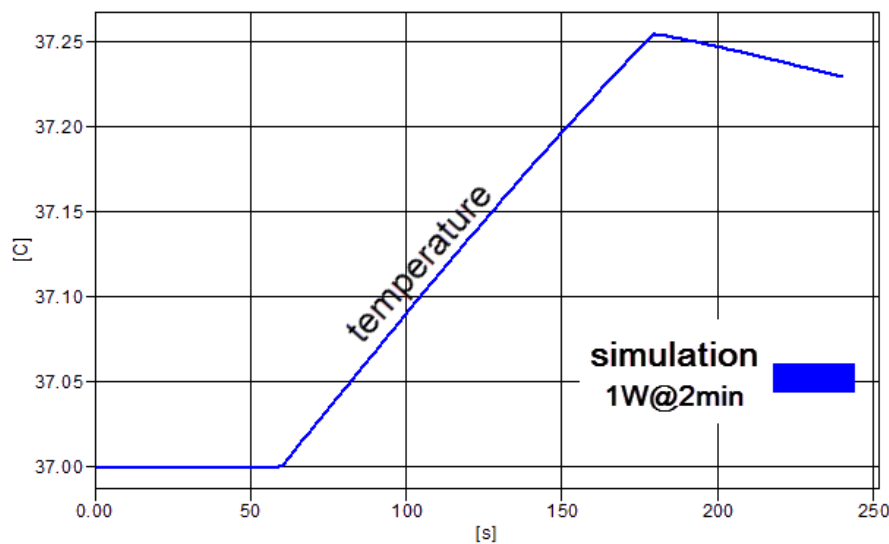


Figure 5.40: Hot spot heating over 2 minute period for 1W transmission.

In this experiment the transmission power varied from 1 W to maximum power, 7.8 Watt continuous wave (CW), and there was an increase in steps of 1 W generating 7 measurements in total. The simulation was constructed similarly with the virtual

Measurement of the hot spot for 1W Tx	Start T	After 2 min	After 4 min	SAR [W/kg]
Measurement 1	26.6	26.8	27	5
Measurement 2	26.6	26.8	26.9	5

Table 5.12: Temperature measurements of the hot spot under the source. For 1 W power input and 4 minutes in total the average increase for a 2 minute period is 0.2°C, which corresponds to a 5 W/kg SAR as defined in equation (5.11).

P_{in} [Watt]	V_{in} [Volt]	Temp [°C]	2min [°C]	4 min °C	pTx W/10s [W]
1W	70.7	10.2	10.3	10.5	1.173
2W	100	10.5	10.8	11.1	2.419
3W	122.5	11.1	11.5	11.8	3.490
4W	141.5	11.8	12.3	12.8	4.655
5W	158.1	12.7	13.3	13.8	5.802
6W	173.2	13.7	14.3	14.9	6.975
7.8W	197.8	14.7	15.7	16.4	blocked

Table 5.13: For the 7 cases the nominal power is calculated from the voltage input, V_{in} , on a 50Ω resistance. The values of the initial temperature, 2 and 4 minute points are given along with the power measured by TALES. The power at the TALES is higher than that at the probe due to cable attenuation which is taken into account.

temperature sensor placed in an appropriate position in order to match the actual optic probe. In this set of measurements the interest is on how the heating rate changes from increased power input. Table 5.13 outlines the nominal power, voltage input, initial, 2 and 4 minute temperatures as well as the power measured by the system after transmission.

A full outline of the experiment along with the temperature rise and deviation from the simulation is shown in the table in figure 5.41. For each of the 7 cases diagrams in figures 5.42, 5.43, 5.44, 5.45, 5.46, 5.47 and 5.48 illustrate the measured temperature versus the simulation. The average deviation, as calculated in equation in figure 5.41, for all cases is 8.4% which indicates that the simulation and measurement slopes follow the same pattern.

Comparison	P _{in}	2min rise [°C]	4min rise [°C]	Deviation* [%]
Measure	1W	0.1	0.3	±5%
Simulation		0.13	0.25	
Measure	2W	0.3	0.6	±7.5%
Simulation		0.25	0.5	
Measure	3W	0.4	0.7	±1.5%
Simulation		0.38	0.75	
Measure	4W	0.5	1.0	±0.5%
Simulation		0.51	1.0	
Measure	5W	0.6	1.1	±9%
Simulation		0.63	1.25	
Measure	6W	0.6	1.2	±23%
Simulation		0.76	1.5	
Measure	7.8W	1.0	1.7	±12%
Simulation		0.99	1.95	

$$* \text{ Deviation [\%]} = \frac{(M_{2min} + M_{4min}) - (S_{2min} + S_{4min})}{2} \times 100$$

Figure 5.41: Outline of the 7 transmission power cases investigated and corresponding temperature rise from 2 and 4 minute scans and corresponding simulation rise. The absolute temperatures are irrelevant but are kept to room temperature for the start of each measurement. The maximum deviation observed was 23% but the average deviation was 8.4% i.e. lower than a 12% error margin.

Dual Loop Measurements

In this experiment the dual channel coil is used and placed over a beef phantom with the two loops designated as channel 1 and channel 2. The diagrams in figure 5.49 show the temperature measurements in real time under source 1 (blue line) with the corresponding simulation (6min@5.2W@170V + 6min cooling) being overlaid (red line). The first peak corresponds to transmitting only with channel 1, the second peak with channel 2 and the third peak with both channels. As discussed, it is of less concern when higher peaks are calculated in the simulations as they provide extra safety margin through overestimation.

A surface heat distribution comparison took place for this setup with the results being summarised in figure 5.50. In this experiment, both channels were transmitting at full power while the sequence was run twice in a row without cooling for

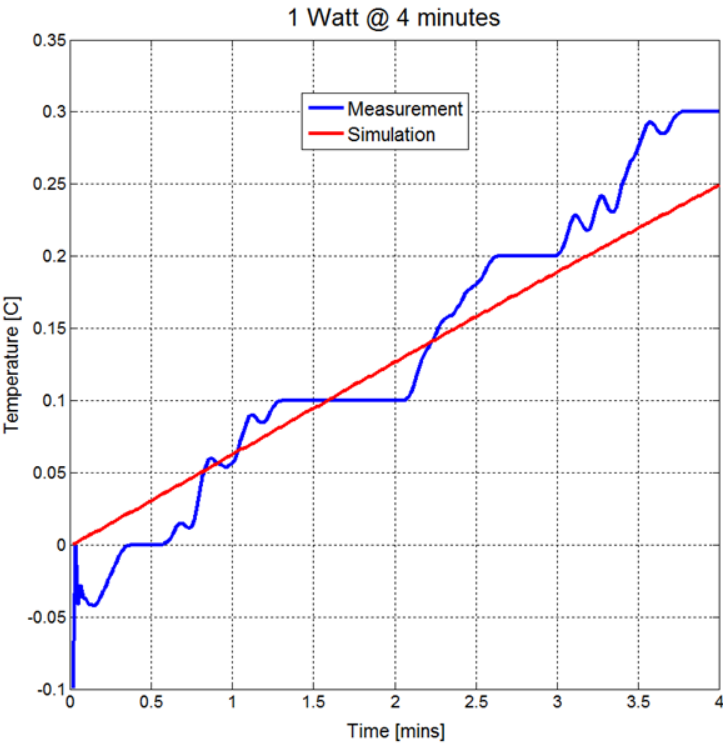


Figure 5.42: Transmission of 1 Watt nominal for a 4 minute scan.

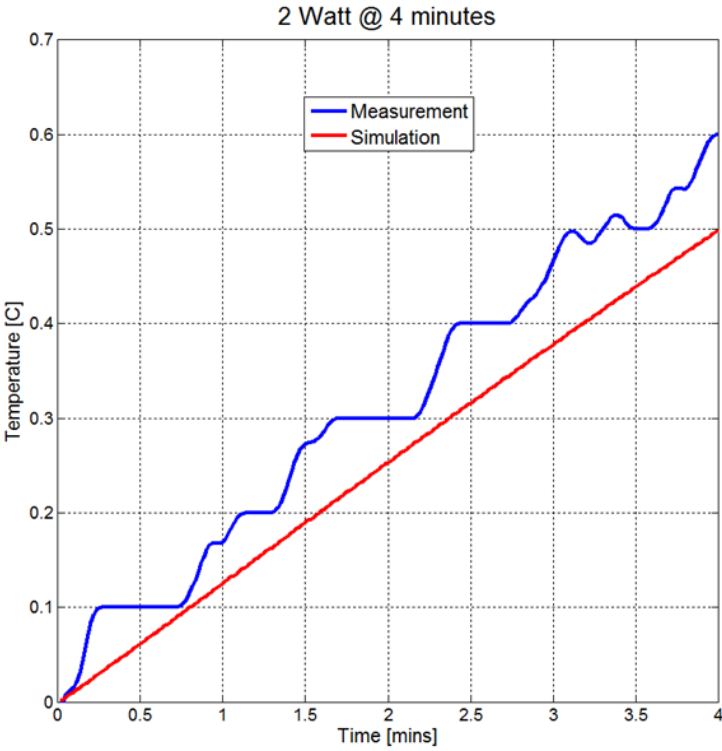


Figure 5.43: Transmission of 2 Watt nominal for a 4 minute scan.

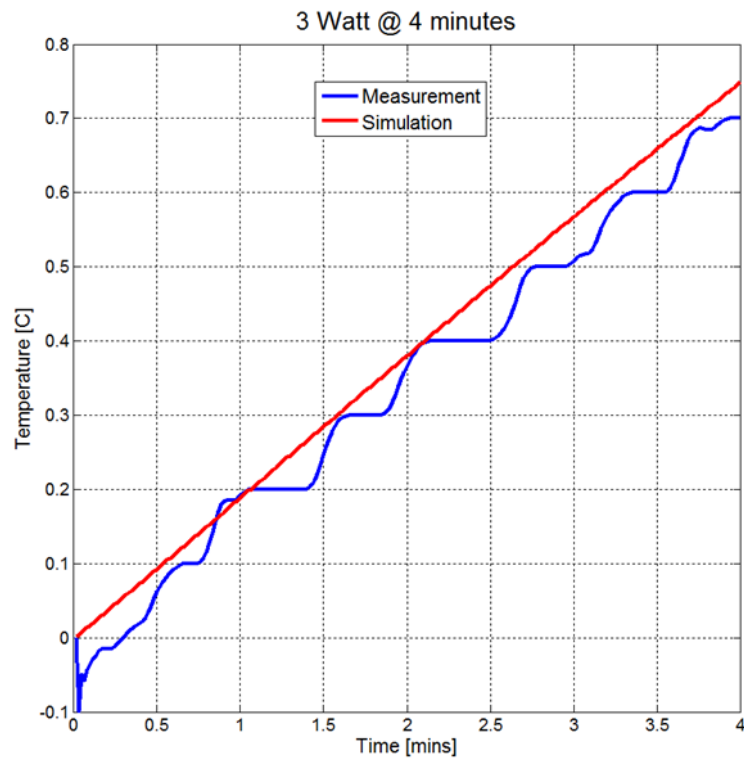


Figure 5.44: Transmission of 3 Watt nominal for a 4 minute scan.

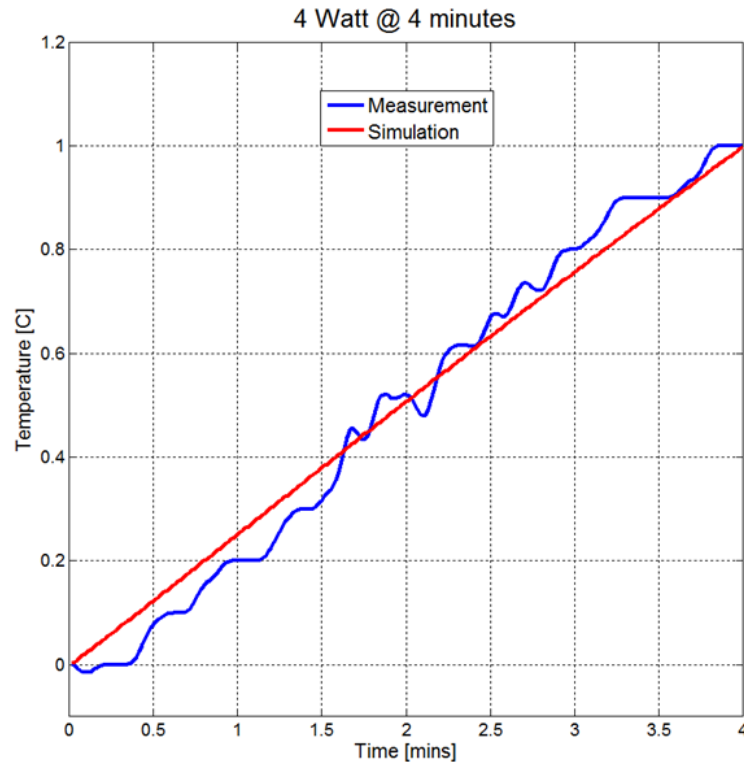


Figure 5.45: Transmission of 4 Watt nominal for a 4 minute scan.

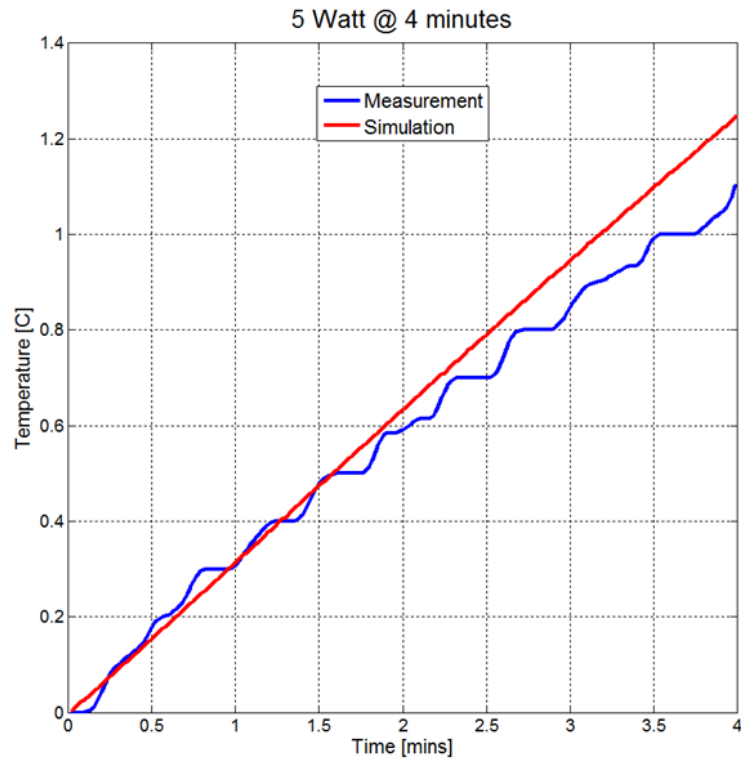


Figure 5.46: Transmission of 5 Watt nominal for a 4 minute scan.

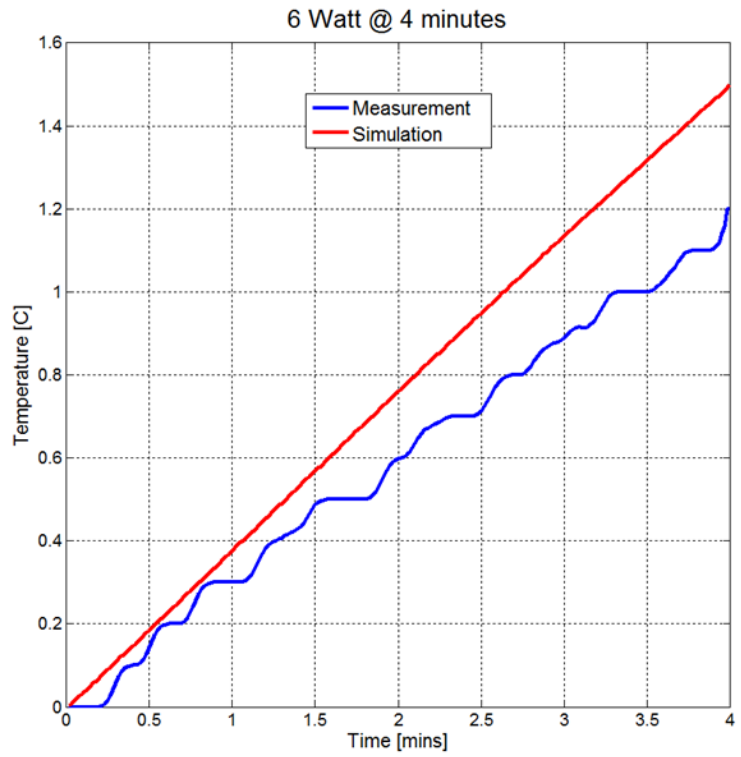


Figure 5.47: Transmission of 6 Watt nominal for a 4 minute scan.

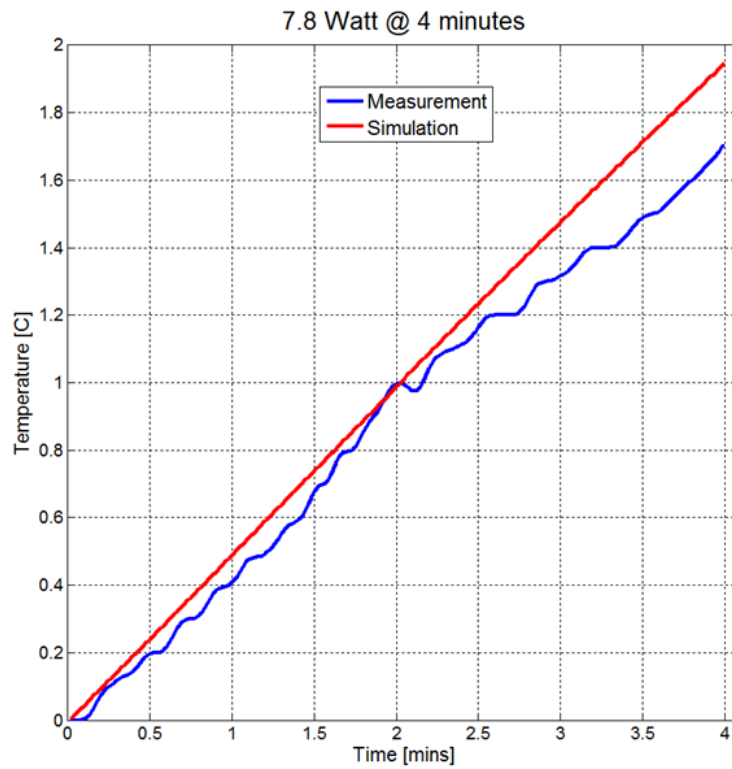


Figure 5.48: Transmission of maximum, 7.8 Watt nominal for a 4 minute scan.

maximum heating. The phantom was taken out in a safe area in order to capture the temperature distribution of the liquid crystal sheet. During the movement of the meat phantom from the scanner room to the computer room for photography it was observed that some of the colouration faded due to the environment temperature. Thus, the contrast in the photo was enhanced to compensate for this effect.

Due to the crystal sheet colour vs temperature logarithmic profile, figure 5.34, the surface heat distribution in simulation is presented in a logarithmic scale in order to make comparison more intuitive. The uneven heating distribution can be attributed to the unbalanced positioning of the coil relative to the phantom.

5.6.3 Discussion

The results show that the temperature rise measured during the scans was very close to the simulated case. A small variation was observed but it was found to be an

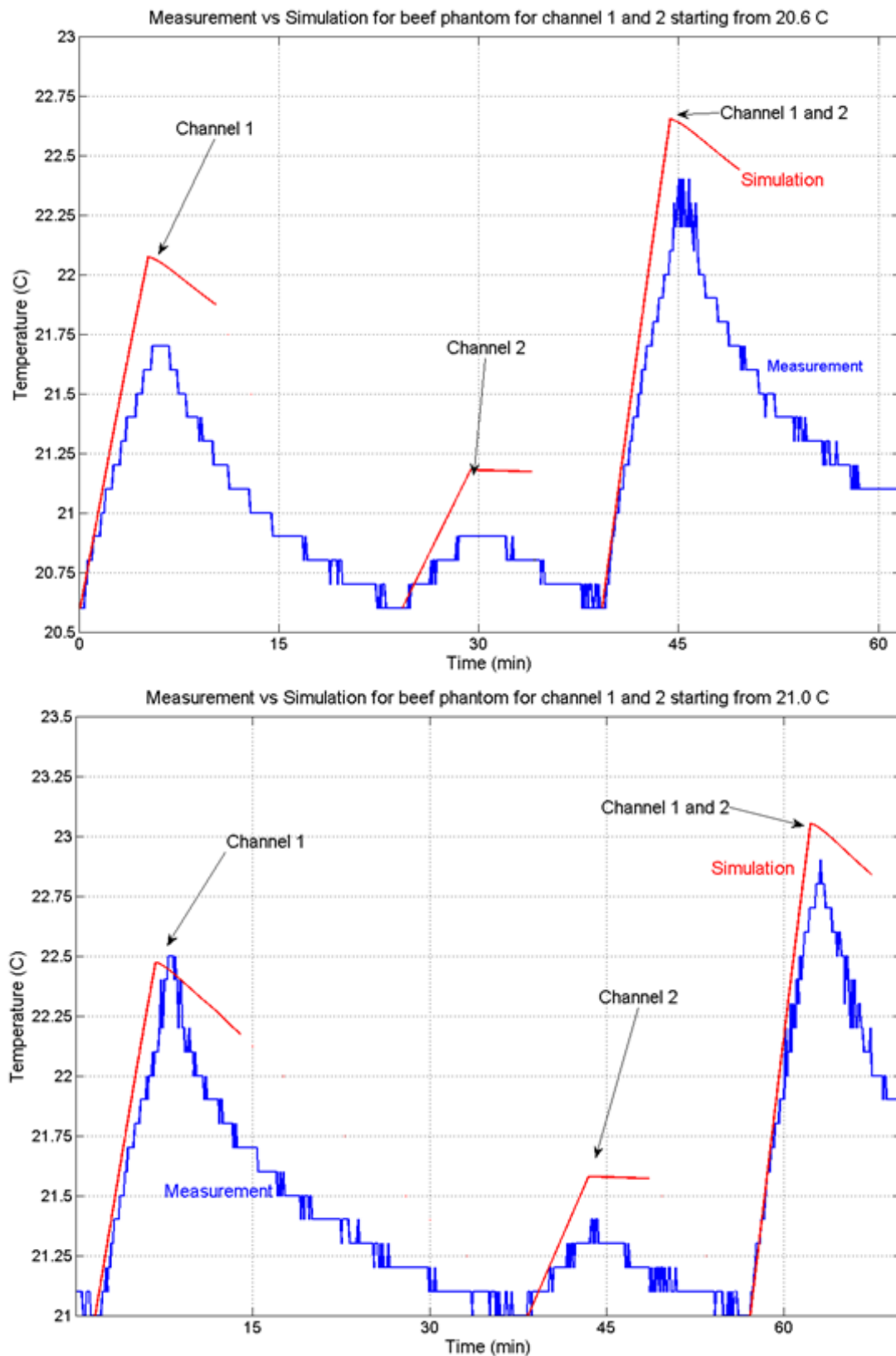


Figure 5.49: Two similar measurements with different initial temperatures. The blue line represents the measurement and hence is shown in discrete steps due to the temperature device resolution. The blue line is continuous because the phantom is left to cool off so that the next measurement remains unaffected by the previous. The red line represents the simulation that is run under the same conditions and left to cool off for 6 minutes. The slopes are the most important aspect since the heating deposition rate is the crucial factor and they match almost perfectly.

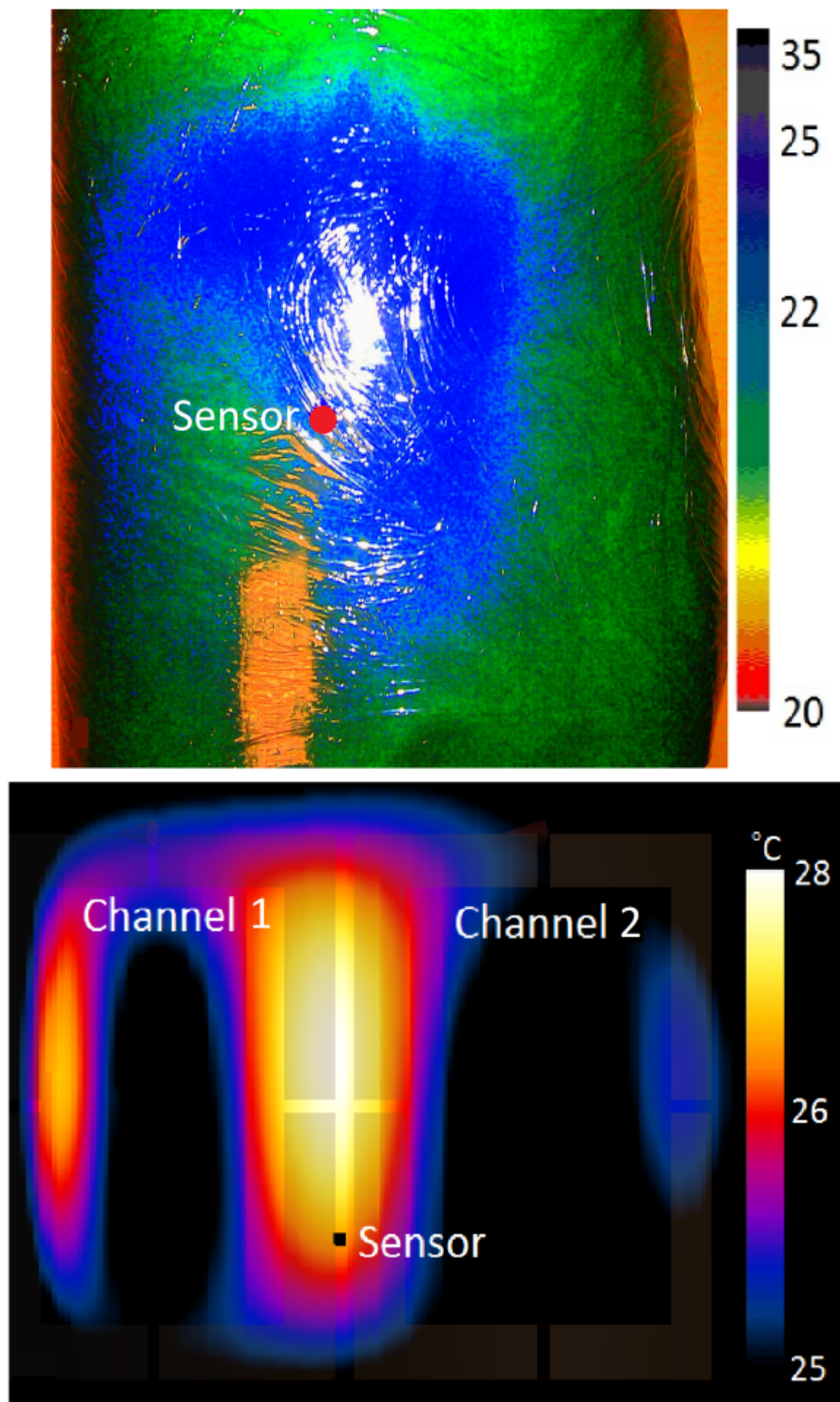


Figure 5.50: The upper panel shows the photograph taken just after a max power scan. The image contrast was enhanced in order to accentuate the heating profile. The lower image shows the simulation under same conditions. The two heating profiles approximate each other reasonably well.

overestimation of the simulation which constitutes an additional safety margin and ensures that the limits are never exceeded in the real case. For visual comparison of the simulation and measurements, liquid crystal sheets were used to observe the heat distribution after the scan.

The results detailed in this section are considered sufficient to assert that the simulation package generates accurate field estimations. Hence, the safety parameters calculated in the previous section via the use of the estimated $\mathbf{E}$ -fields are regarded accurate and safe.

5.7 Conclusion

A comprehensive outline of the physics governing electromagnetic heating and RF safety has been provided. Following this, the regulatory limits set by the accredited international electrotechnical committee were discussed and the RF supervision system was presented thoroughly. The safety validation of the proposed and constructed neck 4Tx/4Rx coil followed outlining the steps taken to ensure accurate simulation conditions. The local SAR supervision is controlled by the framework of virtual observation points that in effect is the clustering of large computational domains to just a few points that can be easily monitored in real time. Finally the simulation was validated with experimental temperature measurements.

The next step after coil validation is subject scanning and evaluation of the coil's performance with respect to image quality and SNR as well as effectiveness in suppressing the flowing blood signal. The coil's performance will be evaluated by measuring quality parameters of both the receive array, e.g. noise correlation matrix, g-factors, and the transmit array, e.g. isolation, and reflection coefficient etc.

Chapter 6

Coil Performance Evaluation

A suitable coil has been designed, constructed and its safety has been evaluated and validated. The next step is to evaluate its performance as an RF and imaging device.

6.1 Simulation and Hardware Evaluation

6.1.1 Simulation Results

A finite difference time domain software package, Semcad-X (Speag, Zurich), was used for the CAD construction of the coil and its evaluation. The transmit channels were simulated with the receive array present and its components set at the values they have in transmission mode. The PIN diodes in particular behave like resistors with a value of $0.3\ \Omega$ when conducting, the passive decoupling circuit being set to resonate at the Larmor frequency with the added inductor. The resonance shift between the simulation with and without the receive array was found to be less than 1%, i.e. 3 MHz, for the worst case.

Once the particular model was setup, the coil was tuned to the correct Larmor frequency with 3 to 4 iterations where the values of the lumped elements were adjusted. The resonance and decoupling between the adjacent elements in both Tx

and Rx arrays was evaluated. The Tx array tuning per element is shown in figure 6.1 and the coupling between its elements is given in figure 6.2. The same data are provided for the receive array in figures 6.3 for the resonances and figure 6.4 for the coupling of the elements.

6.1.2 B_1 -to-SAR ratio

The receive array, however, was found to cause a certain degree of perturbation in the field from the transmit array. In particular, figure 6.5 shows how the B_1^+ value was affected by the presence of the array. The highest B -field decrease was 29.3% with the average among the two phantoms and the two channels being 25.7% relative to the field when the array is not present. The $35\mu m$ thick, $25\times 35mm$ copper plates, used as a ground plane for the preamps, are thought to have had an additional impact in the field reduction and thus they were included in the simulation. The other components, such as the twisted DC cables, the unshielded BALUN traps in the input and output of the preamps, as well as the cable itself were not included in the simulation.

Along with the B -field however, the E -field was also found to be affected in the same proportion. This effect in fact had some beneficial implications in the setting of the necessary power limits. The worst case 10g SAR was found to be reduced by 27%, when the array was included in the simulation, while the averaged B_1^+ reduction, as seen in figure 6.5, is 25.7%. The fact that the worst case reduction is slightly smaller than that of the B_1^+ shows that the same B_1^+ value can be achieved with a smaller SAR. The latter could have implications in the power requirements as well as in the component tolerance, but if these requirements can be met the small increase in the B_1^+ -to-SAR ratio can be translated to a scan time reduction or to a higher achievable flip angle. A comparative map of the B_1^+ with and without the array is shown in figure 6.6.

The VOP supervision scheme was also investigated with regards to the presence

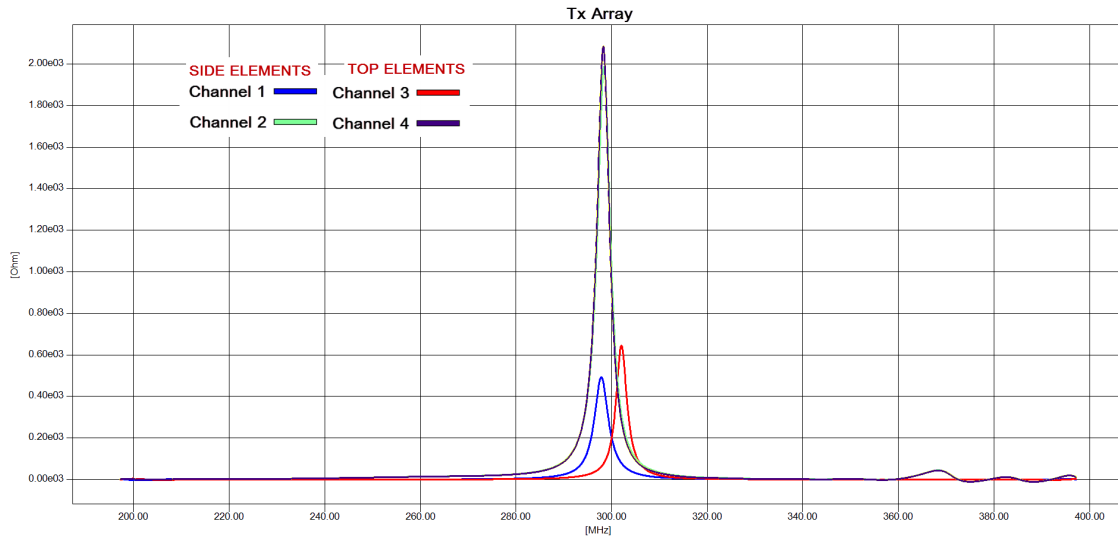


Figure 6.1: Real part of the Tx array element simulation with the inclusion of one of the human phantoms. The two top coils have a different profile due to different position. The iterative process of tuning was followed until each coil in the electromagnetic presence of the rest was resonating at the Larmor frequency.

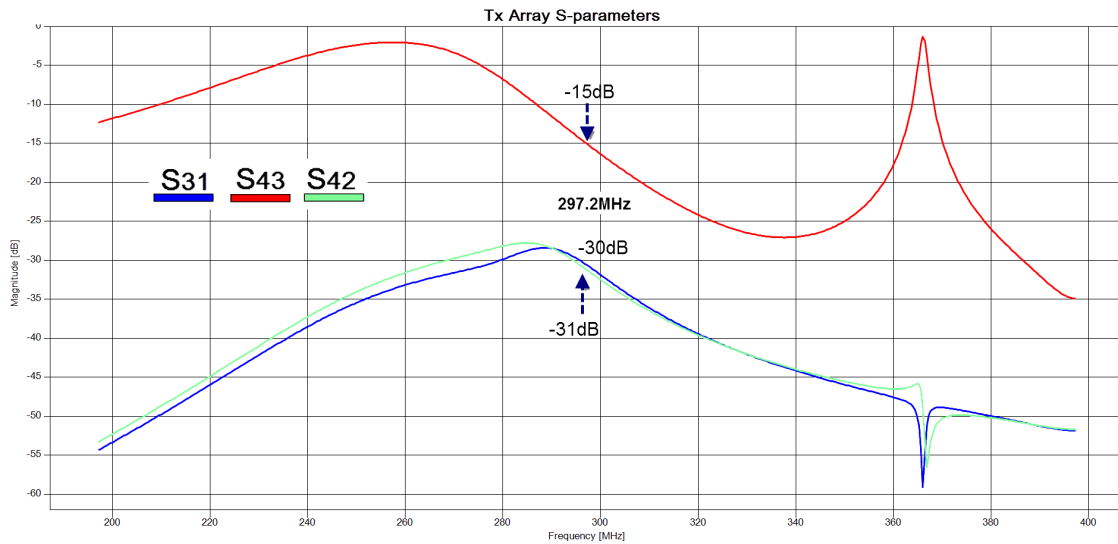


Figure 6.2: Coupling between the Tx elements. The two top coils are geometrically decoupled with -15 dB isolation. S_{12} , S_{14} and S_{23} are excluded because they refer to elements that are physically far from each other and hence exhibit very low coupling.

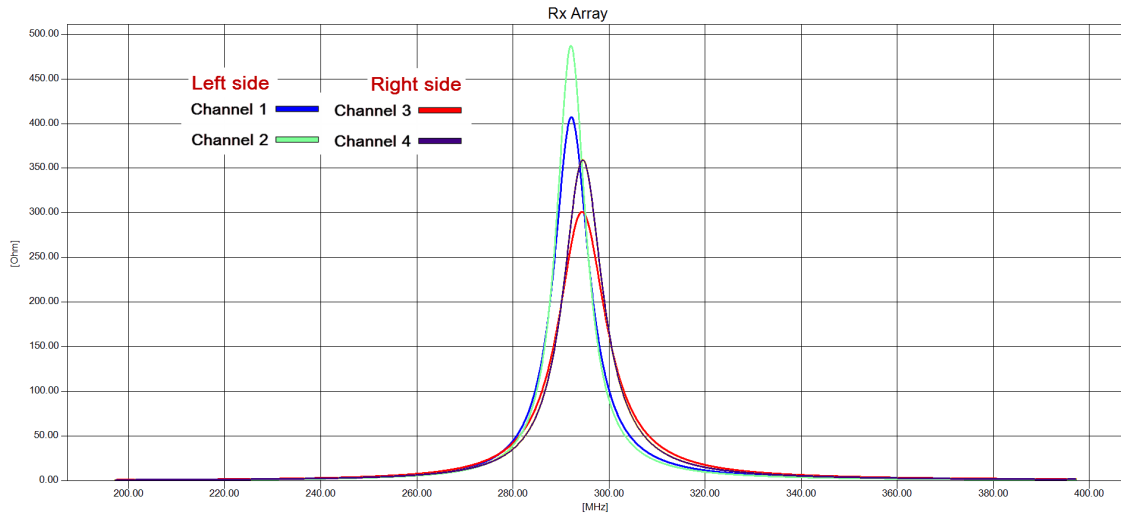


Figure 6.3: Resonances of the Rx elements of the simulation. All coils resonate within a range of 6 MHz around the Larmor frequency. Symmetry of the values of the Rx arrays was kept with the discrepancies arising from the realistic phantom and imbalanced positioning in the simulation.

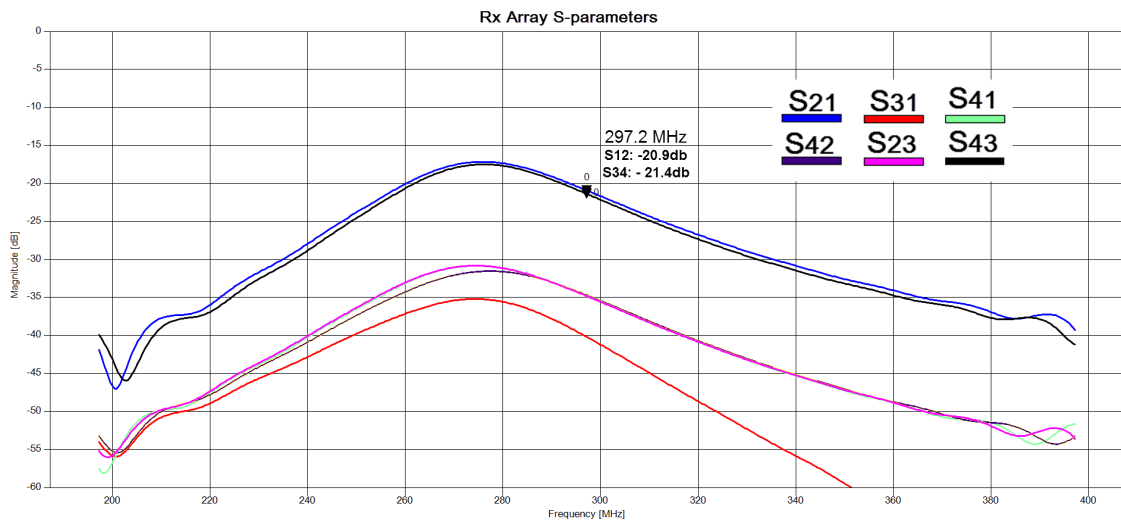


Figure 6.4: Scattering parameters vs frequency of the receive elements. The most interesting values are the S_{12} and S_{34} that show the coupling between the two elements at each side. With appropriate positioning the elements were decoupled to -20 dB or better. This process is crucial to minimise coil interaction and increase the equivalent coil resistance from the coupling induced currents.

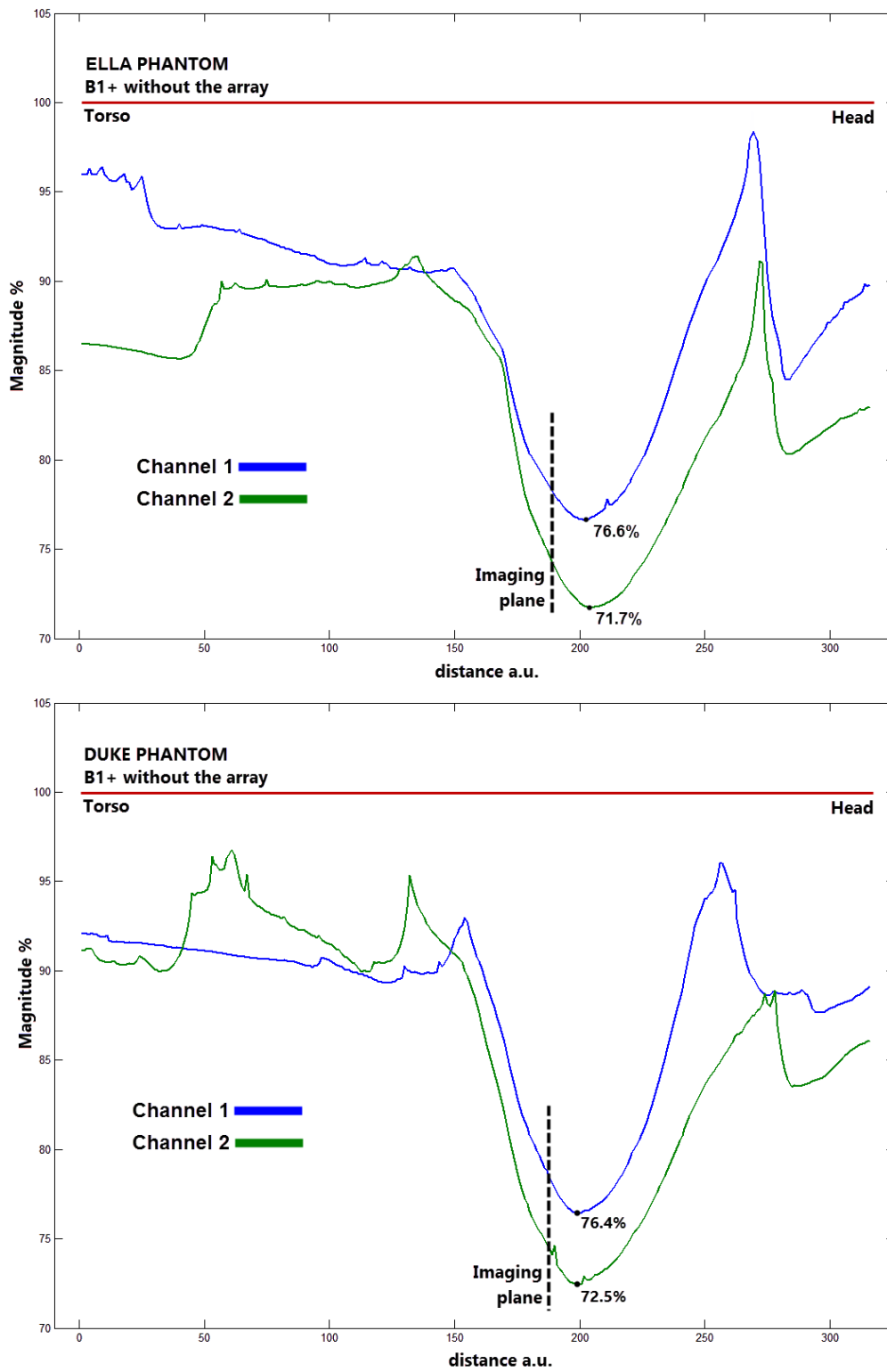


Figure 6.5: The diagram shows how the B_1^+ field is affected around the imaging plane, i.e. where the array coil is positioned. The female (top) and male (bottom) phantom results are presented. The red line indicates the field estimated for no plate. The blue and green lines indicate the averaged field reduction over the transverse plane and the black dotted line the approximate position of the imaging plane.

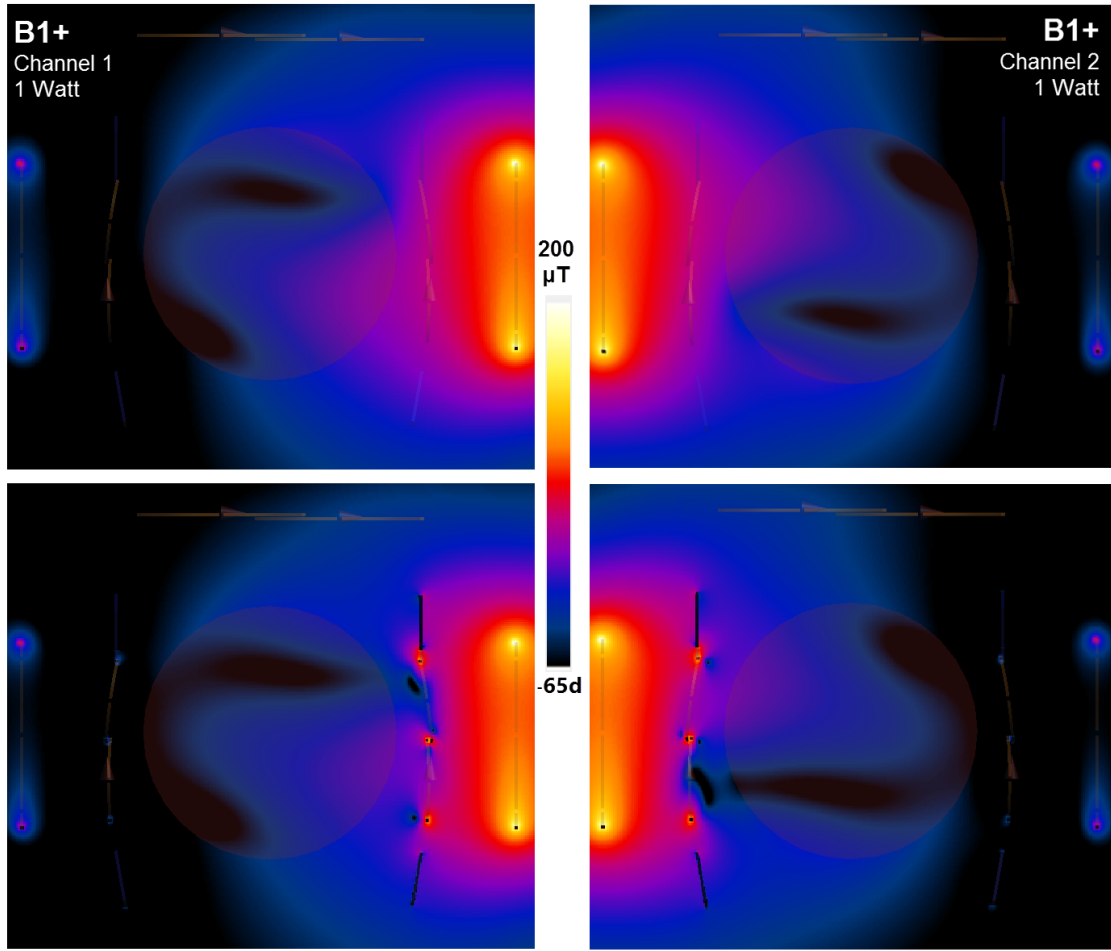


Figure 6.6: B_1^+ drop and perturbation from the presence of the receive array. Both side channels are shown in the left and right columns and the top and bottom rows include the field without and with the array, respectively.

of the receive array along with the preamp plates. The worst case scenario that corresponds to the worst case shim in the VOP was found to decrease from 4.522 to 3.017 W/kg. This 33.3% reduction must be offset with the aforementioned 25.7% B_1^+ drop leading to an overall benefit of B_1^+ efficiency of 11.4%. Table 6.1 provides a quantification of the findings in relation to the effect that the receive array has on the B_1^+ value for the side Tx coils and the overall SAR.

The findings as discussed above were validated with temperature simulation and measurements on the expected hot spot, in front of the top capacitor in the transmit coil, on a meat phantom. A 10 W sequence was run for 3 minutes in each case and the temperature increase under experiment conditions was compared with the

Ella	No plate	Plate	Difference %
Worst Case SAR 10g [W/kg]	7.25	5.24	27.7%
B_1^+ channel 1	1	0.766	
B_1^+ channel 2	1	0.717	
Average B_1^+	1	0.7415	-25.9%

Duke	No plate	Plate	Difference %
Worst Case SAR 10g [W/kg]	7.03	5.13	27.0
B_1^+ channel 1	1	0.764	
B_1^+ channel 2	1	0.725	
Average B_1^+	1	0.7445	-25.6%

VOP worst case [W/kg]	4.522	3.017	33.3%
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SAR efficiency	No plate	Plate	Difference %
B_1^+ /WorstSAR Ella	13.79	14.15	2.6%
B_1^+ /WorstSAR Duke	14.22	14.51	2.0%
B_{1avg}^+ /VOP	22.11	24.63	11.4%

Summary			
Overall SAR efficiency improvement based on Worst Case SAR			2.3%
Overall SAR efficiency improvement based on VOPs			11.40%

Table 6.1: Comparative presentation of SAR efficiency change when the receive array acts as a block to the $\mathbf{B}_1$ field. The findings regarding the drop in the $\mathbf{B}_1^+$ are shown as diagrams in figure 6.5. The SAR efficiency improvement depends on the SAR supervision scheme used. For the use of a worst case scenario as discussed in Chapter 5 the SAR efficiency increases by 2.3%. If the VOP supervision scheme is used, also discussed in Chapter 5, the benefit is variable with an overall minimum of 11.4% indicating that this configuration can be useful.

simulated trace, see figure 6.7. The simulation gave an average 50% drop in the expected temperature increase case. The experimental measurements showed an even greater drop of 70% but, as discussed in the previous chapter, there is a limitation to the accuracy of the measurement due to the low resolution of the temperature measurement system.

6.1.3 Bench Measurements

The coil was evaluated on the bench loaded with a cylindrical water phantom of 120 mm diameter and a Hewlett Packard 8712-C network analyser. The coil was

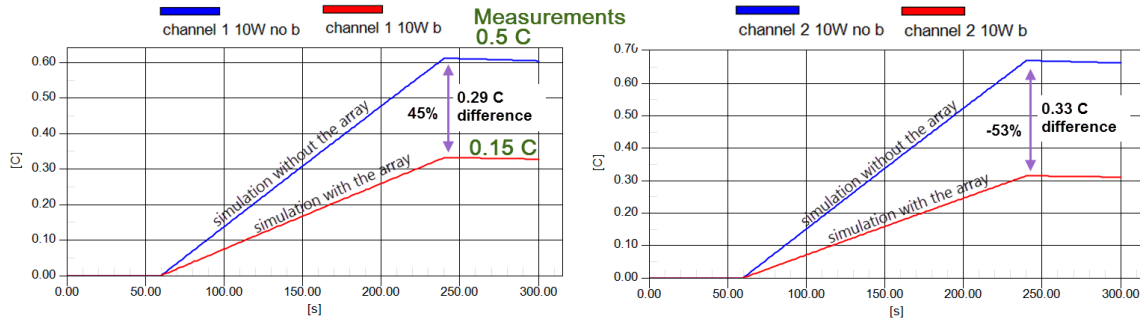


Figure 6.7: Temperature increase simulation on a meat phantom for left (channel 1) and right (channel 2) channels on the hot spot. The blue line represents the heating for a 10 W CW transmission for 3 minutes without the Rx array between the phantom and the Tx element. The red line shows the temperature increase under the same transmission parameters but with the presence of the receive array and its components between the transmit element and the phantom. In green, the average from two measurements on the hot spot of a meat phantom are summarised. The simulated reduction in power deposition is 50% and 70% in the measurement, indicating an improvement in the SAR profile of the coil regardless of how the $\mathbf{B}_1$ is affected.

treated as an 8-port system, one port for each element and after cable calibration the scattering parameters were measured for three modes of operation. Table 6.2 summarises the scattering parameters for the coil when it was not biased by any external DC voltage. Table 6.3 shows the S-parameters when the coil operated in transmission mode, i.e. all Rx array PIN diodes were ON. The receive elements were terminated with 50Ω. Finally, table 6.4 outlines the S values for the coil operating in receive mode. The Tx coil was this time actively detuned. The measurement between Tx1 and Tx2 is omitted since the side Tx elements are positioned far from each other and interact very little.

An important measurement for the coil is the unloaded Q-factor to loaded Q-factor ratio. The load, as discussed in Chapter 4, will introduce additional losses that can be seen as a resistance added to the intrinsic conductor resistance. The magnetic losses indicate that the subject is magnetically coupled with the coil but there is no way to differentiate where most of the Q-factor drop comes from. However, if balancing has been implemented on the probes, the dielectric losses will be minimised. Table 6.5 shows how the receive loops are affected by the load distance.

Unbiased 50Ω terminated - 0.5 gal phantom								
	Rx1	Rx2	Rx3	Rx4	Tx1	Tx2	Tx3	Tx4
Rx1	-6dB							
Rx2	-6.8dB	-17.2dB						
Rx3	-28.5dB	-31dB	-8.8dB					
Rx4	-25dB	-25.4dB	-8.3dB	-13.8dB				
Tx1	-9.5dB	-16.4dB	-30dB	-33.8dB	-3.4dB			
Tx2	-31.5dB	-31.8dB	-14dB	-12.2dB	X	-4dB		
Tx3	-19.4dB	-21.7dB	-22.5dB	-22.2dB	-20dB	-31.5dB	-9.2dB	
Tx4	-39.4dB	-23.6dB	-20dB	-21.3dB	-32dB	-19.5dB	-9.6dB	-17dB

Table 6.2: For this measurement none of the PIN diodes was biased. The transmission between all possible pairs of ports when the others are 50Ω terminated was measured. The coil was loaded with a cylindrical 0.5 imperial gallon phantom of water. S-parameter between Tx1 Tx2 is omitted since the elements, being far apart, interact very little.

Biased in Tx 50Ω terminated - 0.5 gal phantom								
	Rx1	Rx2	Rx3	Rx4	Tx1	Tx2	Tx3	Tx4
Rx1	-0.5dB							
Rx2	-27.2dB	-0.1dB						
Rx3	-52dB	-50dB	-0.1dB					
Rx4	-48.5dB	-44.1dB	-40dB	-0.1dB				
Tx1	-16.6dB	-17.2dB	-36.4dB	-41.7dB	-25dB			
Tx2	-40.3dB	-53dB	-25.3dB	-22dB	X	-15dB		
Tx3	-25.3dB	-24.3dB	-36.2dB	-37.8dB	-13.3dB	-26dB	-15.9dB	
Tx4	-37.5dB	-33.6dB	-35.7dB	-34.2dB	-33.9dB	-18dB	-9.9dB	-24dB

Table 6.3: For this measurement the PIN diodes of the receive array were biased and the scattering parameters were measured. The coil was loaded with a cylindrical 0.5 imperial gallon phantom. In bold is the geometric decoupling of the two top coils.

The actual neck-coil distance can vary depending on the subject but an average distance is found to be 20 mm. In table 6.6 the Q_U/Q_L ratio is given for each loop for a load placed at 20 mm. The average ratio for all loops is found to be 3.4. Finally, the water phantom used was selected to have a similar loading effect as the human tissue.

6.1.4 Scanner Measurements

The receive array was tested by calculating the noise correlation coefficient matrix by simulation and by data collected during scans without RF excitation for each

Biased in Rx 50Ω - 0.5 gal phantom terminated								
	Rx1	Rx2	Rx3	Rx4	Tx1	Tx2	Tx3	Tx4
Rx1	-22.4db							
Rx2	-13dB							
Rx3	-24.6dB	-30.2dB	-36.3dB					
Rx4	-17.7dB	-25.7dB	-13.5dB					
Tx1	-22.1dB	-26.4dB	-42.5dB	-38.9dB	-1.3dB			
Tx2	-38.2dB	-36.1dB	-26.7dB	-29.5dB	X			
Tx3	-27.8dB	-31.5dB	-29dB	-38dB	-40dB	-54dB	-0.6dB	
Tx4	-47.4dB	-29.1dB	-22.8dB	-33dB	-46.5dB	-39.5dB	-26.5dB	

Table 6.4: For this measurement the PIN diodes of the transmit array were biased and the scattering parameters were measured. The coil was loaded with a cylindrical 0.5 imperial gallon phantom. In bold is the geometric decoupling of the two receive loops in each side.

Load at	5mm	10mm	15mm	20mm	30mm
QL	17	36	47	79	230
QU	271				

Table 6.5: Q-factor measurement for receive channel 1 for different loading positioning at a distance of 20 mm from the phantom. The closer the load the more loading resistance it introduced.

channel. The noise covariance maps from each channel were correlated and normalised based on [175, 176, 177]. The resulting matrix is shown in figure 6.9. The simulation shows higher correlation between the neighbouring elements as expected. The experimental results confirm the simulation finding, showing similar correlation between the two elements at each side and very little interaction between the loops from different sides.

For parallel imaging performance the geometric factor (g-factor) was estimated to further evaluate the coil. For this calculation, the root sum of square SNR image was used to calculate the noise amplification for $R=2$ and $R=2 \times 2$ [111]. The FOV was selected at 120 mm to fit on the diameter of the cylindrical phantom and restrict additional aliasing outside of the selected phantom. The simulation and experiment are compared in figure 6.10 where the inverse noise amplification map is shown. Comparison of the two maps shows that the coil can in practice bear an R value of 2 or less but for two dimensional reduction the noise is too high and imaging becomes

	Channel 1	Channel 2	Channel 3	Channel 4
QL	79	85	86	82
QU	271	290	265	302
QU/QL	3.43	3.41	3.08	3.68

Table 6.6: The ratio of the Q_U/Q_L is presented for each of the Rx elements. Each receive loop was constructed to be identical to the others so the average ratio was 3.4 with a standard deviation of 0.2.

impractical.

Figure 6.8 shows the simulated and measured root sum-of-squares (RSS) reconstruction from the Rx array. The phantom was the same and RF shimming on a circular ROI in the centre of the bottle was performed before the measurement. Nevertheless, the $\mathbf{B}_1$ remained considerably inhomogeneous relative to the ideal distributed field and generated the asymmetrical image shown in the figure.

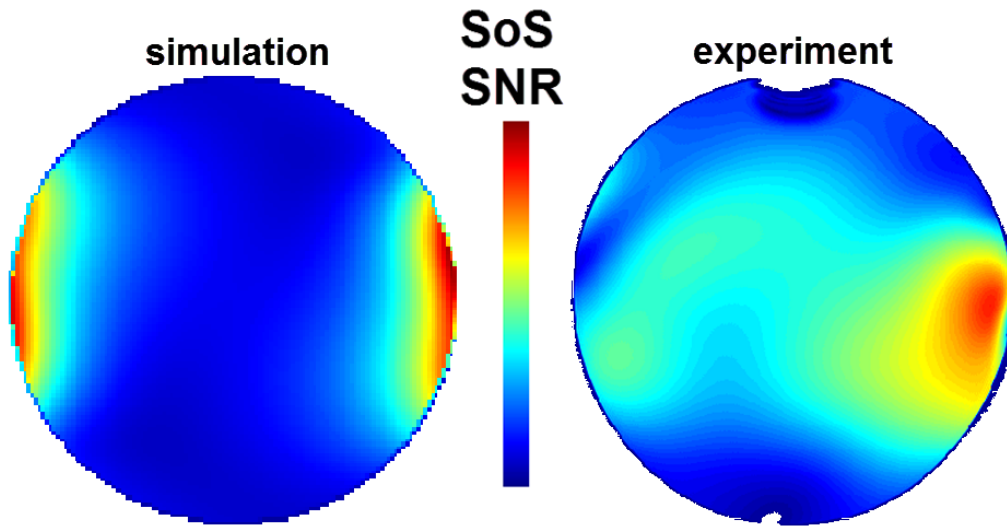


Figure 6.8: Sum of squares magnitude optimal reconstruction. The image on the left illustrates the simulation and on the right the experimental map. For the scan, $\mathbf{B}_1$ shimming was performed for a central circular ROI but the inhomogeneous $\mathbf{B}_1^+$ field generated the observed asymmetry.

The coil was first used in the scanner with the previous phantom and its $\mathbf{B}_1^+$ field was compared per channel with that estimated from the simulation. Figures 6.11 and 6.12 show the simulated $\mathbf{B}_1^+$ field on the left against the scanned $\mathbf{B}_1$ mapping on the right. The $\mathbf{B}_1$ mapping sequence used was based on a turbo FLASH sequence

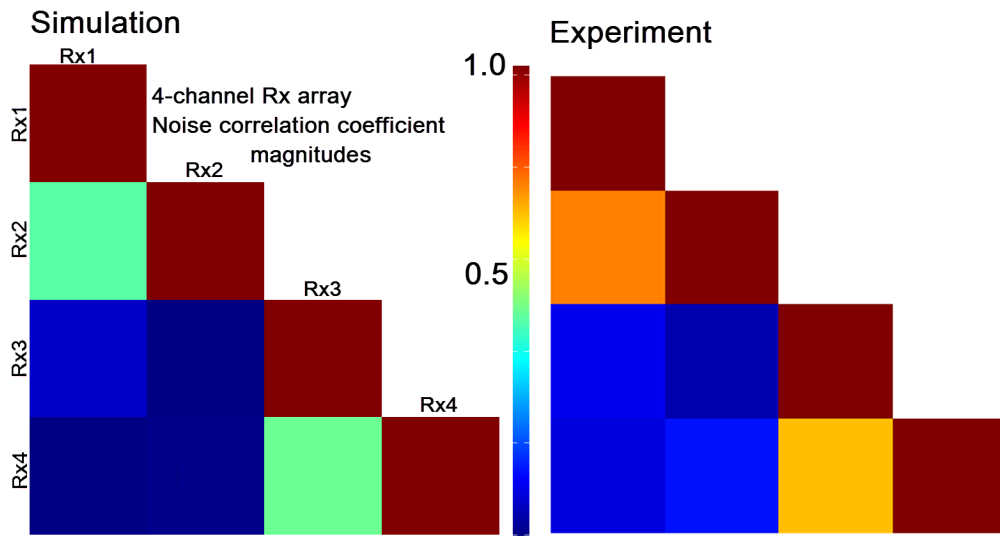


Figure 6.9: Channel noise correlation coefficient matrix (magnitudes) for the 4 channel Rx array. On the left the simulation is shown and on the right the calculation from noise maps acquired with no RF excitation. Matrices approximate each other well. For the right matrix, a utility sequence was used with permission from [178].

developed by Siemens (Erlangen, Germany) for the 7 T system. Flip angle = 15° ; resolution = $0.8 \times 0.8 \text{ mm}$ and 5 mm slice; FOV = 200 mm; TE/TR=3.68 / 9 ms. Each channel transmitted individually while all the receive channels were active.

6.2 Flow Suppression Evaluation

6.2.1 Phantoms

The achieved blood signal suppression with the use of the coil was evaluated with the use of a flow phantom comprising a 10 mm diameter flexible plastic tube pumping water via a peristaltic pump. For the coil loading, a beef meat phantom of 2.0 kg was used. The overview of the experiment along with an imaging plane slice is shown in figure 6.13. The selected fluid velocities were 10, 20, 30 and 40 cm/s in order to include the statistical maxima and minima of the blood velocities in the carotid arteries. The coil was set to operate within safety power limits and managed to achieve the desired flip angles by adjusting the TR as required. The experiment was repeated for inverse flow in order to assess the suppression in both directions.

The results are summarised per fluid velocity selected in figures 6.14, 6.15, 6.16 and 6.17.

The suppression was evaluated by measuring the reduction ratio of the averaged signal in the lumen (green circle) between the two flow directions. If A is the signal from the unsuppressed fluid and B is the signal from the inverse and thus suppressed fluid the reduction is calculated as $((A - B)/A) \cdot 100\%$. The calculation was designed so that the DANTE train could be kept in both measurements and hence factored out with regard to its effect on the static tissue signal. The average suppression for 10 cm/s was found to be 78.8%, for 20 cm/s it was 80.3%, for 30 cm/s it was 77.3% and for 40 cm/s it was 73.9%. The total average suppression was 77.6% which is deemed adequate. The results per experiment and per velocity are summarised in table 6.7.

10cm/s	unsup-pressed	sup-pressed	Reduction		30cm/s	unsup-pressed	sup-pressed	Reduction
1	97.61	10.68	89.1%		1	83.51	13.42	83.9%
2	115.37	26.29	77.2%		2	103.48	22.29	78.5%
3	234.14	56.21	76.0%		3	223.36	48.5	78.3%
4	208.39	56.28	73.0%		4	180.08	56.45	68.7%
Average			78.8%		Average			77.3%
20cm/s	unsup-pressed	sup-pressed	Reduction		40cm/s	unsup-pressed	sup-pressed	Reduction
1	76.56	7.8	89.8%		1	80.1	10.34	87.1%
2	105.38	19.87	81.1%		2	78.83	18.6	76.4%
3	206.27	50.17	75.7%		3	113.01	56.18	50.3%
4	228.14	57.96	74.6%		4	224.28	40.37	82.0%
Average			80.3%		Average			73.9%

Table 6.7: Summary of the fluid signal suppression in the flow phantom. The numbering refers to the tubes as marked in figure 6.14. The average suppression is calculated for each case. The signal is normalised in a range from 0 to 256.

6.2.2 Subject Scans

MR images from consenting healthy volunteers, who satisfied the screening procedure, were obtained for coil evaluation. A fast low angle shot (FLASH) read-out was

used for a 2D and 3D reconstruction. The parameters of the DANTE module and the RO section are given in table 6.8. The pulse width was increased from $60\mu m$ to $200\mu m$ to compensate for the restrictive power regime of 7 T. For a 15° flip angle in the carotid arteries, the required $\mathbf{B}_1^+$ was found to be approximately $5\mu T$. The input voltage was calculated off-line and was found to be 55 V rms to achieve a mean of $5\mu T$ along the artery and 87 V rms input to achieve $5\mu T$ for the ‘average - standard deviation’ field in the carotid artery. Both power input levels were used with direct implications in the rate of deposited power and thus the scan duration.

	<i>Parameters</i>		<i>Parameters</i>
DANTE module		FLASH READ OUT 2D/3D	15 degrees flip angle
	150 pulses		0.6mm in plane resolution
	$\tau=200\mu s$ RF pulse		2.5mm/0.6mm slice
	1.5 ms		TE: 3.68 ms / TR: 10/5 s
	inter-pulse spacing		GRAPPA R=2
	15 degrees flip angle		130 Hz/px
	18 mT/m gradient in every direction		256 Base resolution 100% PE steps

Table 6.8: The parameters used for the DANTE prepared FLASH are summarised here. For the realisation of the flip angles the required voltage input for the known pulse was calculated off-line and fed into the coil.

After the calculation of the voltages and the RF shim on the slice, the TR was modified in order to fit within the allowed power limits. The restrictive nature of the SAR supervision scheme selected gave a 3D sequence of 12 min for R=2 which is an unusually long duration in a clinical setting. For the 2D protocol 12 slices were selected and the higher voltage was applied. The total duration of the sequence was 4 minutes. Another 4 minutes of idle scan time were required between the two protocols for the 6 minute averaging to reset and to avoid any scan interruptions for the RF monitoring system.

Anatomical images for the localisation are shown in figure 6.18 with a mean SNR value of 134 for the right internal and external carotid, above the bifurcation. Absolute and relative $\mathbf{B}_1^+$ mapping followed and an RF shim was applied in the

oblique slice on the imaging plane of the coil as in the process shown in figure 6.19 for phantoms. In figure 6.20 PD sagittal images of the cervical spine are shown with the vertebrae visible and marked. The average SNR in the CSF was measured at 92 while in the vertebrae at 80. The SNR was measured based on the standard deviation of a background area in the ROI.

The subject images taken with the DANTE module activated are shown in figure 6.21 for the 2D protocol and in figure 6.22 for the 3D protocol. The SNR is measured for the left external and internal carotid arteries and in the static tissue closer to the skin. The CNR between blood and the artery lumen is also evaluated in both images. Finally, black blood images taken at 3 T with the same sequence parameters and a 4-channel coil are shown in figure 6.23. The results of the comparison between the images with the constructed coil at 7 T and the commercial one at 3 T are discussed in the next chapter.

6.3 Summary

In this section, a systematic coil performance evaluation was executed. In the first part of this chapter the basic performance criteria were simulated. The result of the simulations showed the ideal profile for the designed coil. In comparison with the experimental data there was a variation ranging from a fairly good match such as in the case of the g-factor to apparent discrepancy such as in the case of the SoS SNR. In the case of discrepancy the experimental result method needs to be evaluated more carefully as per the methodology used.

Regardless of the discrepancies in some measurements the image results show that the RF coil provides an adequate $\mathbf{B}_1^+$ coverage for the available power. The duration of the sequence varied from 12 to 4 minutes due to the restrictions in power transmission. The constructed coil, even though it has just 4 receive coils, was successfully used in parallel imaging mode with a reduction factor $R=2$ helping

to realise a feasible scan duration. While an otherwise adequate SNR was achieved, motion and SENSE noise along with the DANTE effect in the static tissue signal degraded the image quality.

The next, final chapter will provide a short discussion about the research done in this thesis. A thorough future work section will follow and the final conclusions are then drawn.

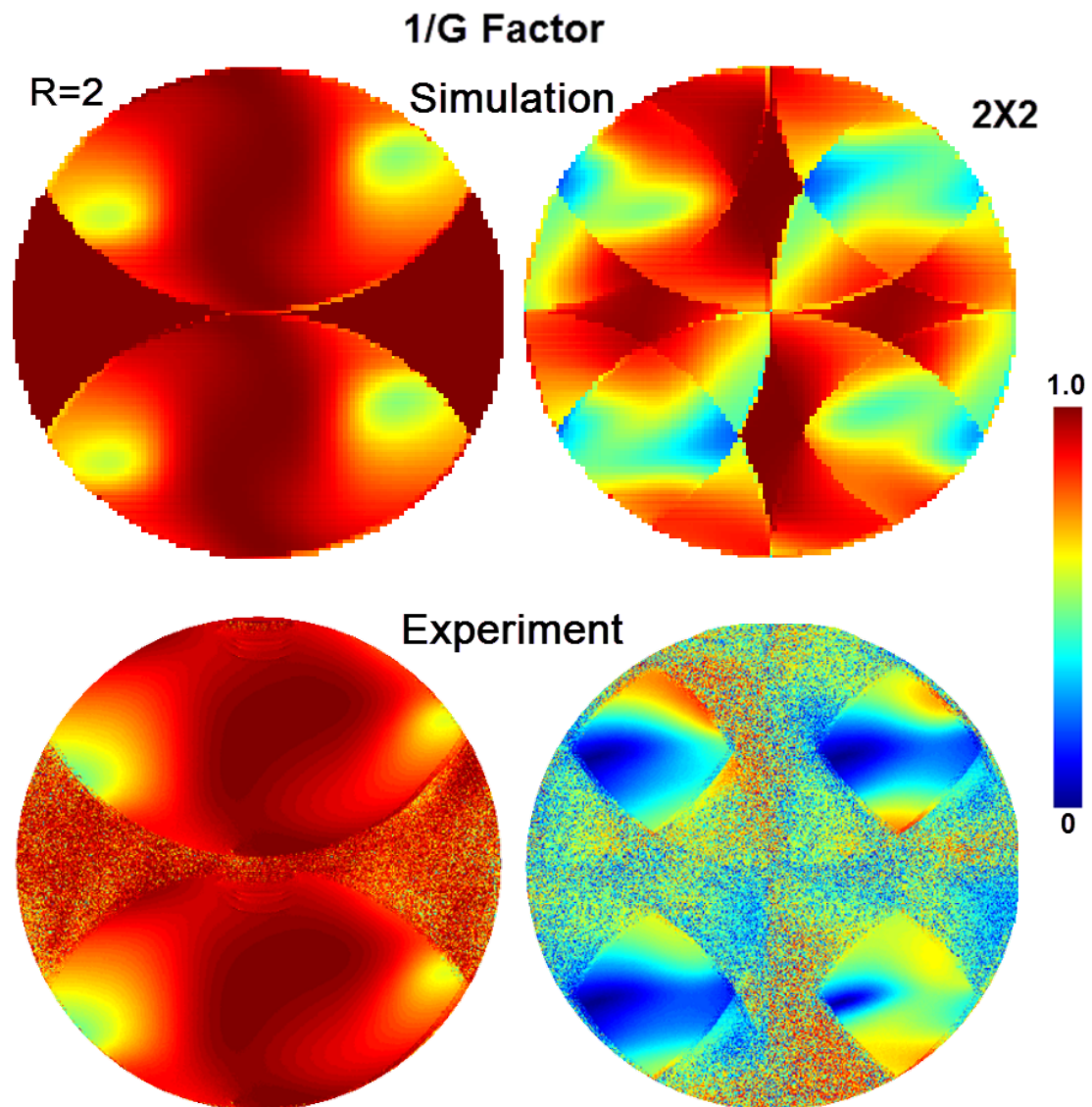


Figure 6.10: Geometric factor or g-factor maps, the SENSE noise amplification by the use of parallel imaging techniques where the k -space is filled based on the sensitivity profiles. The noise is dependent on the geometrical arrangement of the elements. The acceleration factor is noted on the top left of each map. For the bottom row images a coil evaluation sequence was used with permission from [178].

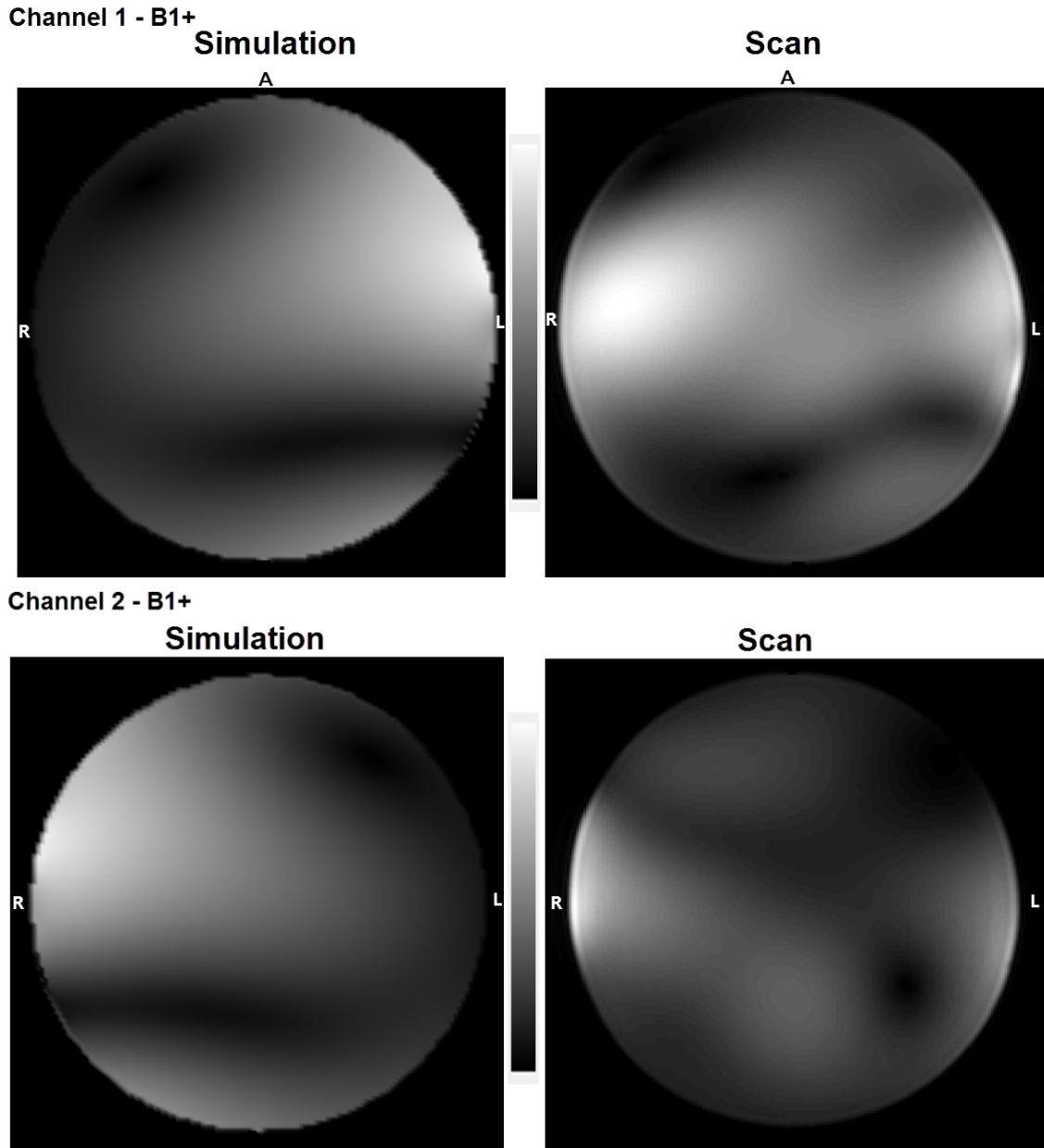


Figure 6.11: Comparison of the simulated B_1^+ versus a FLASH image acquired by the selected transmit element while signal received from the Rx array. On the top row the left side channel is shown and on the bottom row the right side channel is shown. The two fields approximate each other with the discrepancies arising from the receive sensitivity profile as well as the inaccuracy of the coil positioning. The omission of some parts present in the receive array (cables, BALUNs), which are placed between the transmit element and the phantom, can alter the achieved B_1 field too.

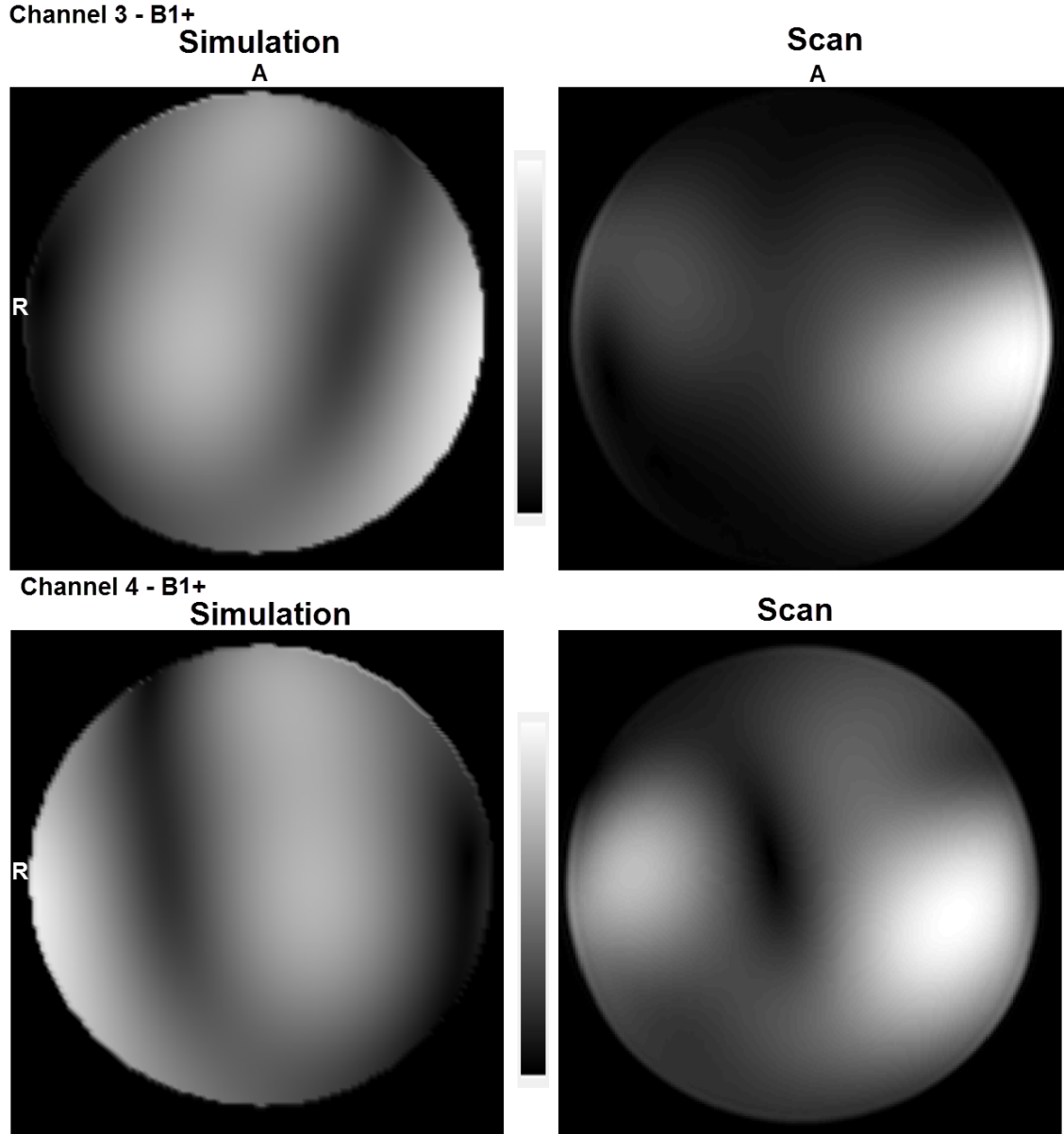


Figure 6.12: Comparison of the simulated B_1^+ versus a FLASH image acquired by the selected transmit element while signal received from the Rx array. On the top row, the left top channel is shown and on the bottom row the right top loop is shown. Due to the imaging plane being further from the top coil position the field is weaker compared to the side coils.

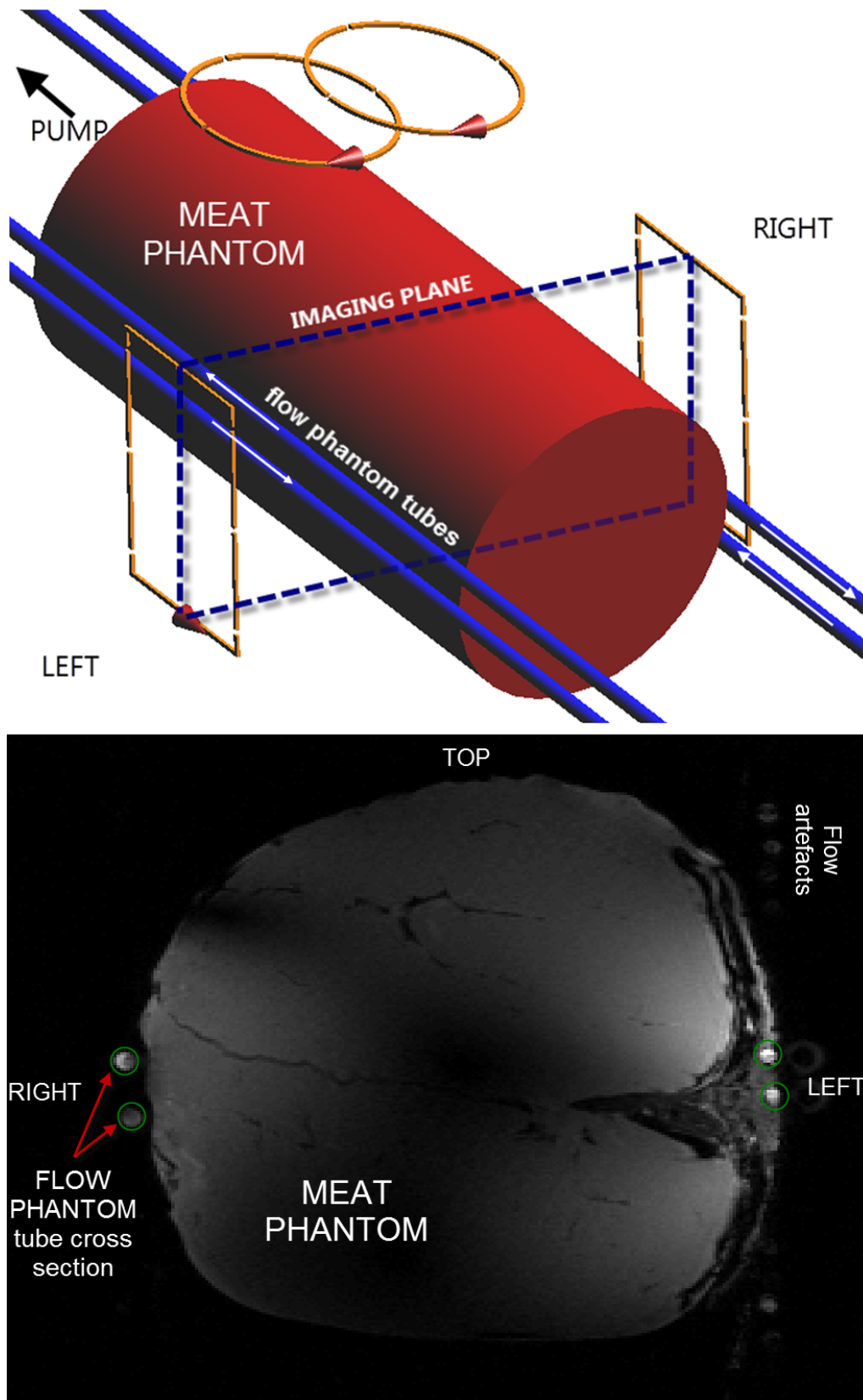


Figure 6.13: Upper panel shows an illustration of the phantom experiment. A peristaltic pump was used with latex tubing of 4 mm diameter. The velocity was calibrated in advance with the use of timing fluid delivery for known volumes. The lower panel shows the transverse imaging plane of the meat and flow phantom taken with a FLASH sequence and no DANTE suppression. Flow artefact is evident on the upper right corner of the image.

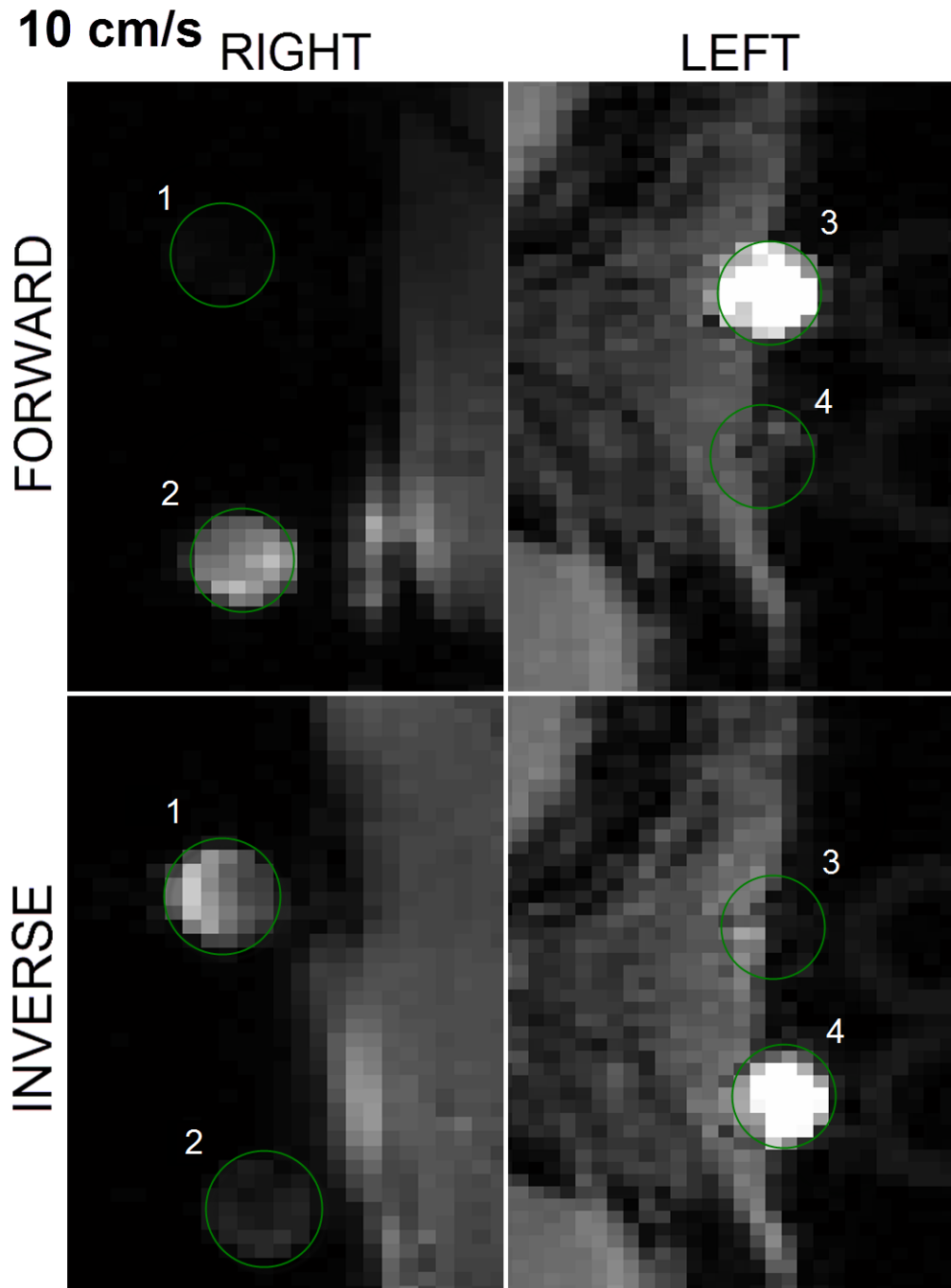


Figure 6.14: Flow suppression with the use of a DANTE preparation module when the flow velocity is 10 cm/s over a 4 mm diameter latex tube. The DANTE pulses were set at $N_p=150$, 15° with 1.5 ms inter-pulse spacing. The gradient along the z-axis was set at 18 mT/m. The read-out was a 128 slice turbo-flash sequence with TR/TE of 25 ms/2.6 ms and $0.6 \times 0.6 \times$ in plane resolution with 0.5 mm slice thickness. The suppression observed in the image was satisfactory for both directions in the imaging plane.

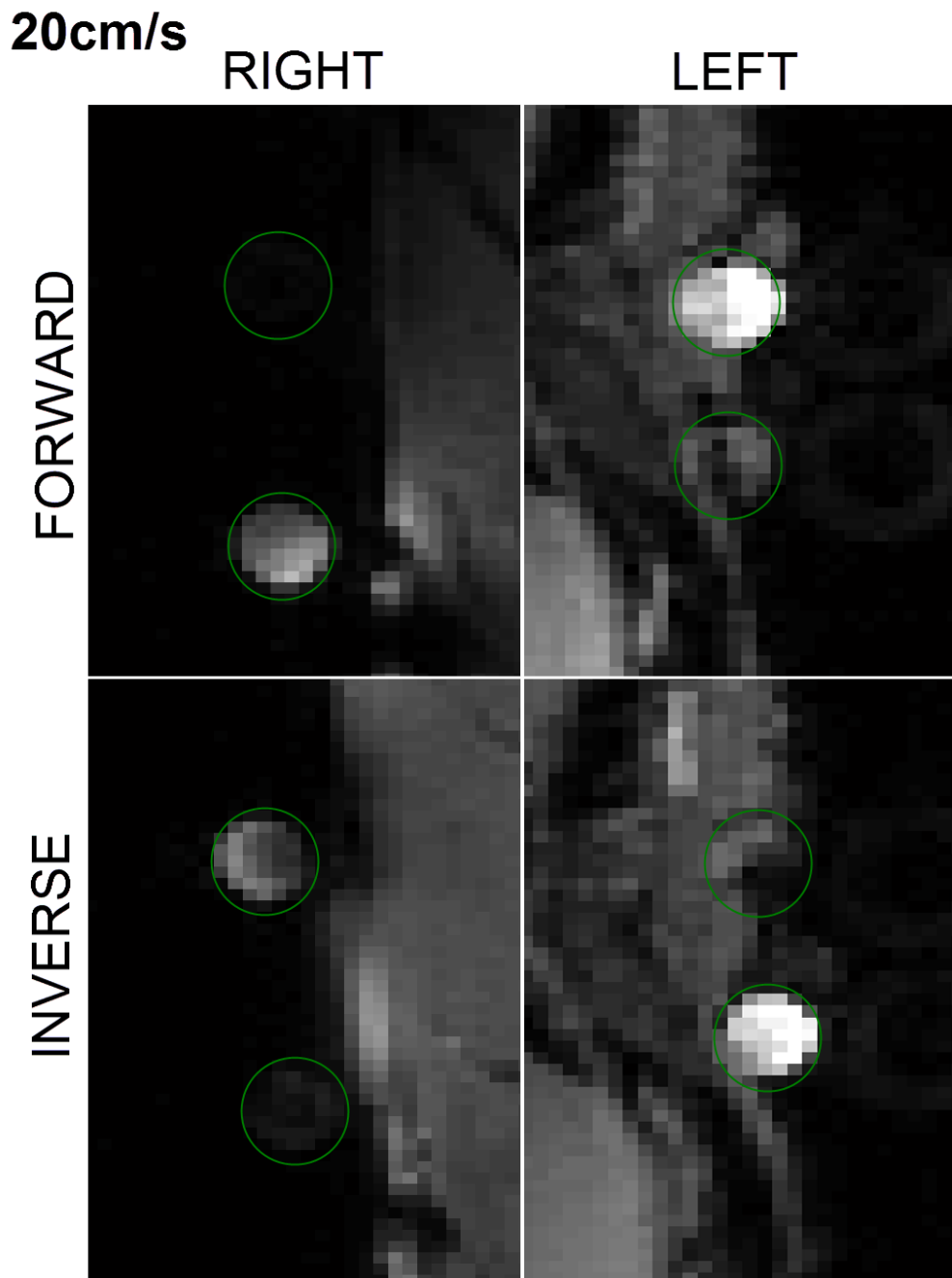


Figure 6.15: Flow suppression with the use of a DANTE preparation module when the flow velocity is 20 cm/s over a 4 mm diameter latex tube. The sequence parameters are as given in figure 6.14. The suppression is satisfactory here too.

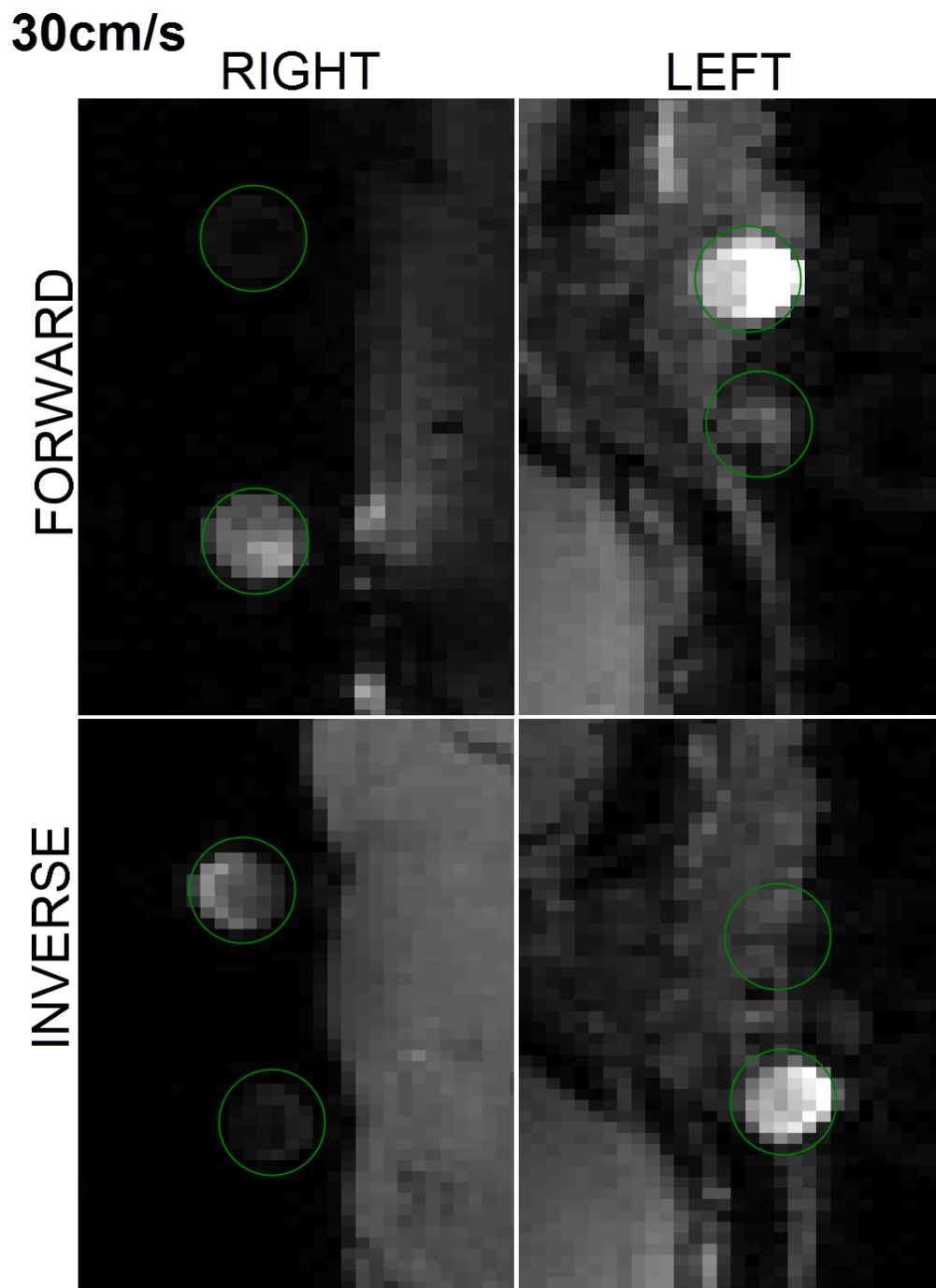


Figure 6.16: Flow suppression with the use of a DANTE preparation module when the flow velocity is 30 cm/s over a 4 mm diameter latex tube. The sequence parameters are as given in figure 6.14. The suppression is satisfactory but not thorough due to the increased flow velocity.

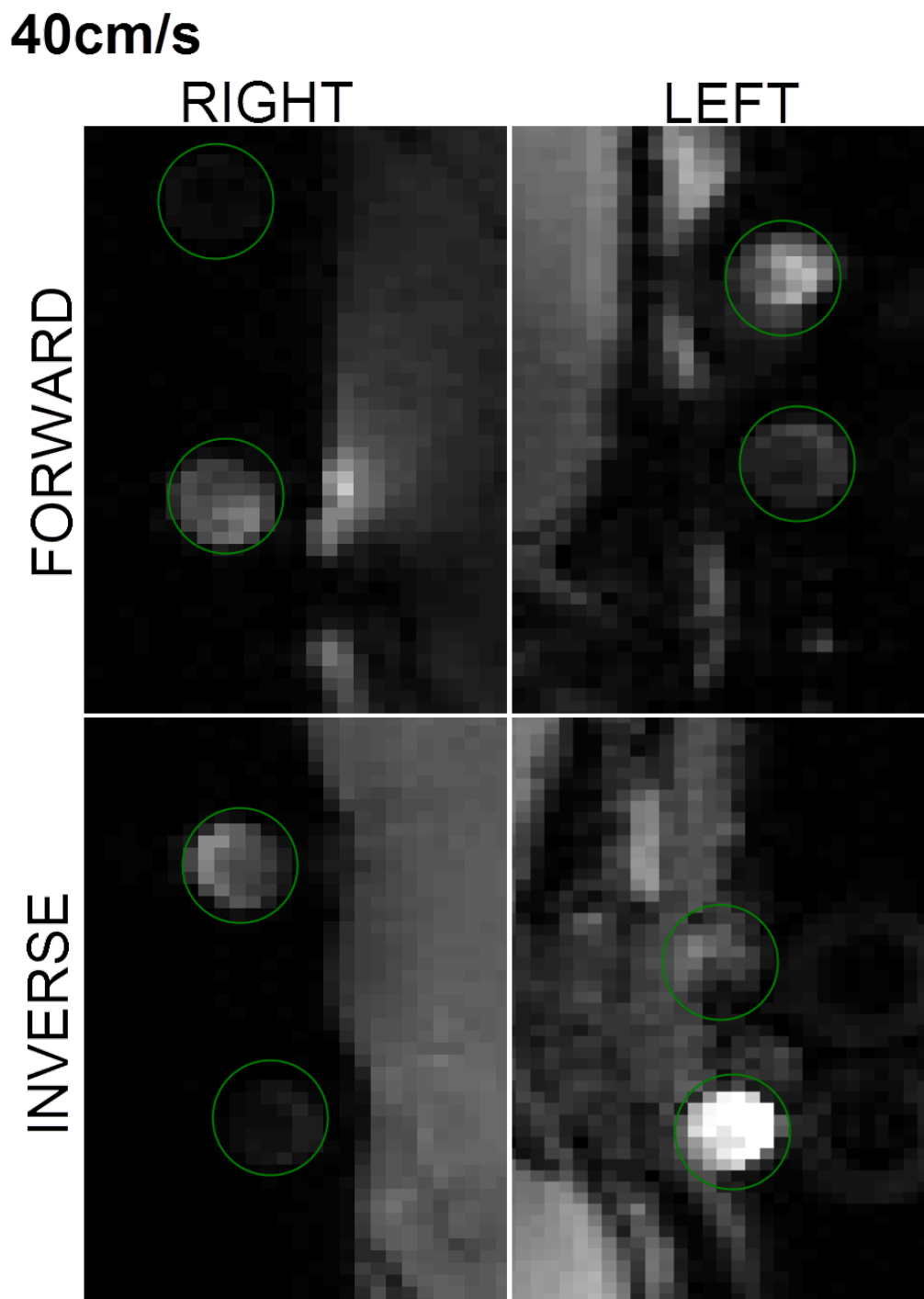


Figure 6.17: Flow suppression with the use of a DANTE preparation module when the flow velocity is 40 cm/s over a 4 mm diameter latex tube. The sequence parameters are as given in figure 6.14. The suppression is satisfactory. This fluid velocity is higher than the one that can be found during diastole in the carotid arteries indicating that the coil can generate satisfactory results in-vivo.

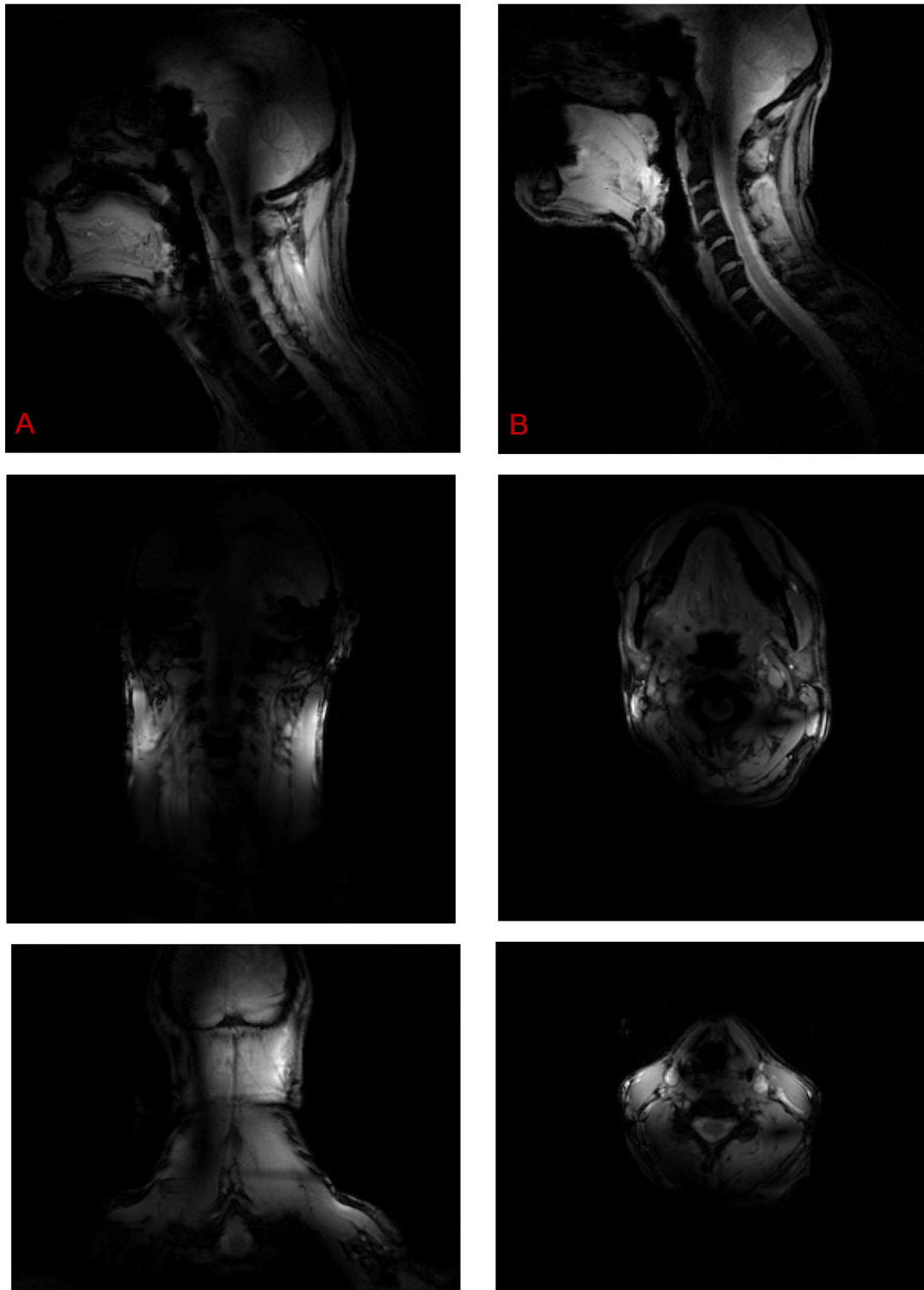


Figure 6.18: From top to bottom, sagittal, coronal and transverse planes of GRE PD scans on two healthy subjects, A and B. Flip angle = 20° ; in plane resolution = 1.2×1.2 mm with 5 mm slice thickness, TE/TR=4/8.6 ms, FOV = 300 mm; base resolution = 256 / 100% PE and BW = 260Hz/pixel.

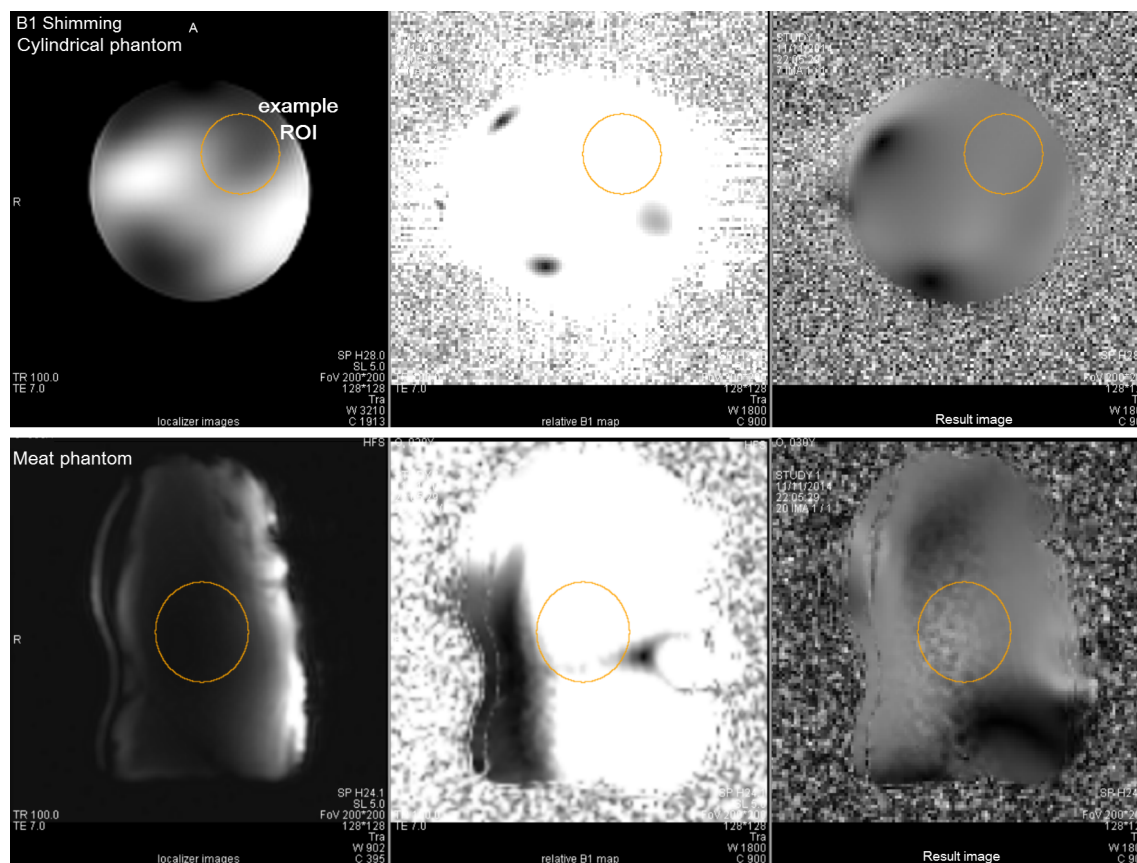


Figure 6.19: B_1 shimming process in the Siemens scanner. Top row, shimming is performed for the cylindrical phantom and on the bottom row for a meat phantom. The left panel shows the anatomical scan where the ROI is selected (circular ROI shown). Middle panel shows the B_1 relative maps and the right panel shows the result image.

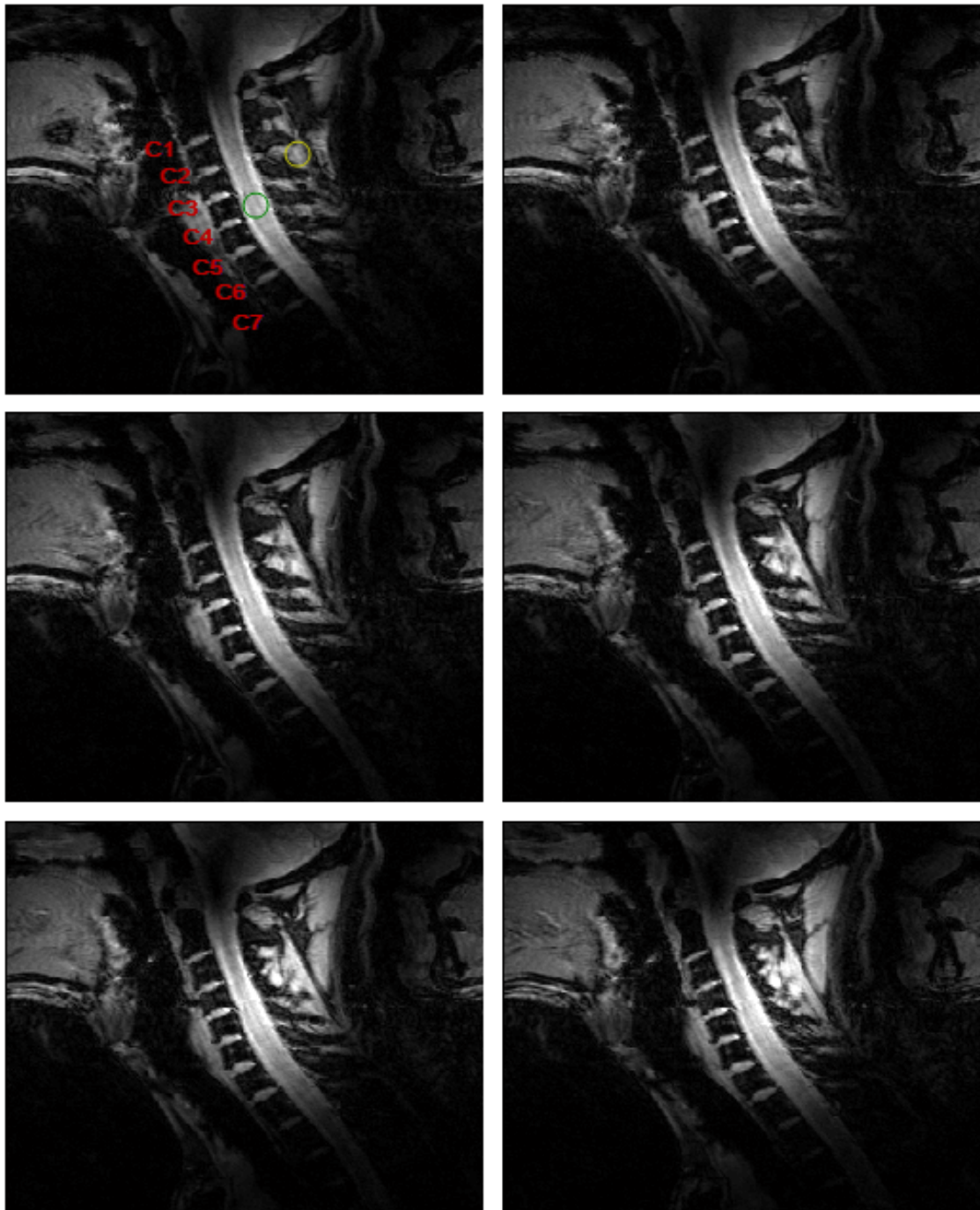


Figure 6.20: PD sagittal imaging of the c-spine. A turbo-FLASH MPR sequence was used with 220×200 mm FOV and 256×256 matrix; slice thickness = 1 mm; TE/TR=3.44/1040 ms; BW = 179 Hz/pixel. The SNR in the spinal cord (green circle) was found to be 92 ± 5 and in the vertebrae (yellow circle) it was found to be slightly lower, at 80 ± 4 .

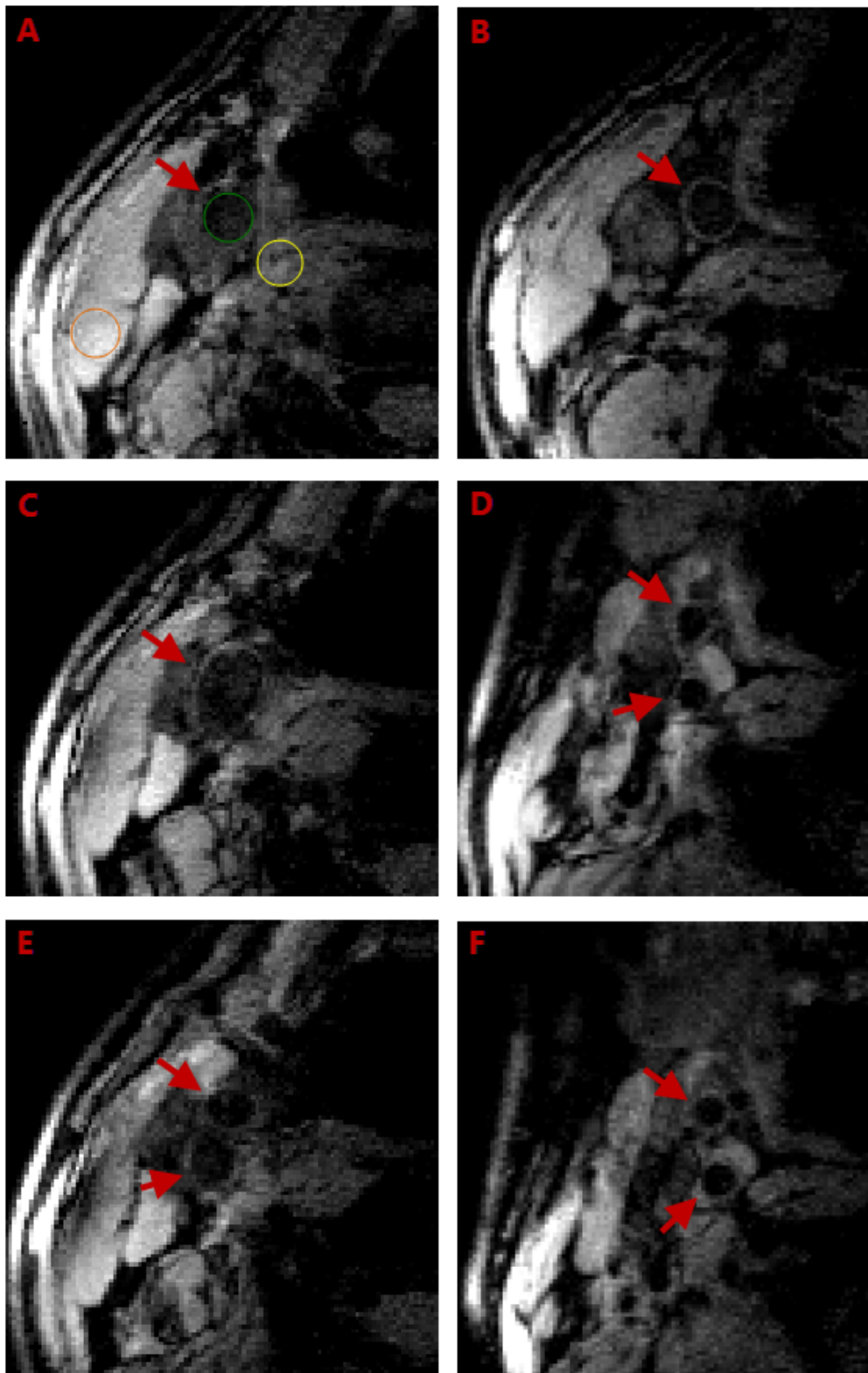


Figure 6.21: Selection of black blood images with DANTE suppression for the 2D protocol. Right carotid artery location is shown by the red arrows. The average measured SNR around the artery (yellow circle) was 24 ± 4 and 56 ± 6 in the static tissue closer to the skin (orange circle). The SNR of the lumen (green circle) was 10.9 and the CNR between the lumen and the tissue was 17.3 ± 2 .

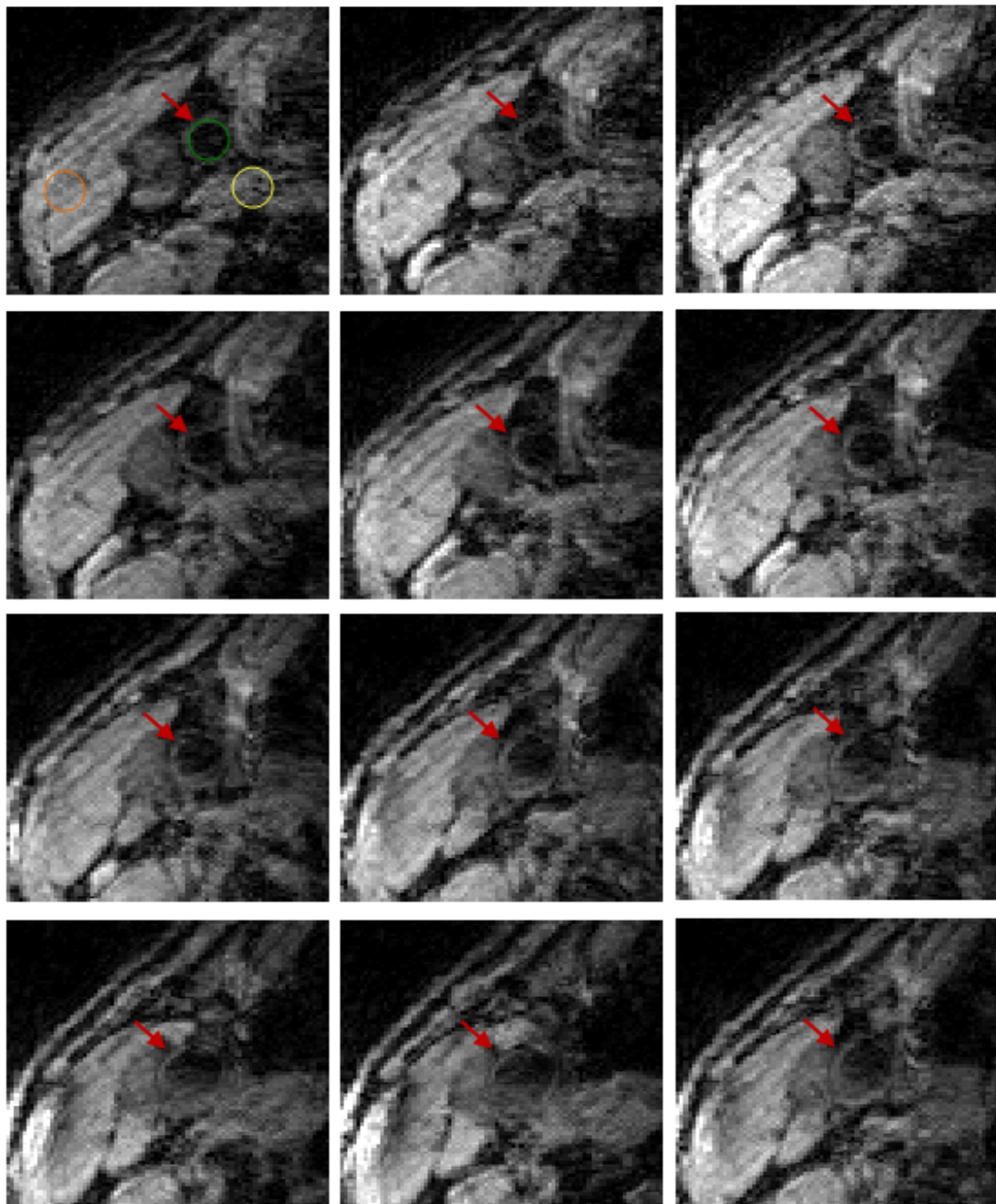


Figure 6.22: Selection of black blood images with DANTE suppression for the 3D protocol. Right carotid artery location is shown by the red arrows. The measured SNR was 33 ± 6 around the artery (yellow circle) and 55 ± 8 in the static tissue closer to the skin (orange circle). The SNR of the lumen (green circle) was 11.5 ± 3.5 and the CNR between lumen and the vessel wall tissue was found to be 24.9 ± 4 showing that there is differentiation between suppressed blood and surrounding tissues.

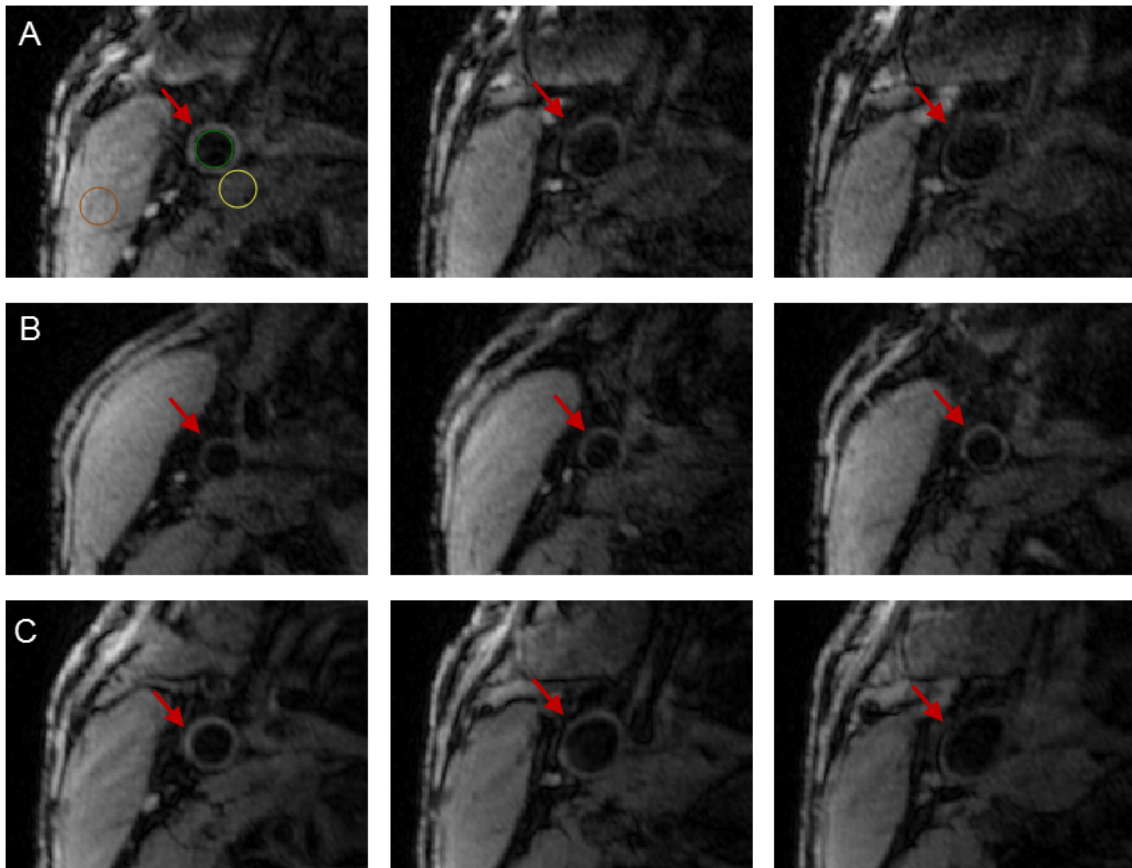


Figure 6.23: The images here are taken in 3 T with the same 3D FLASH sequence and similar DANTE parameters. The resolution is 0.6 mm isotropic, TE: 3.68 ms and DANTE flip angle set to 10° . The SNR for all the images in the static tissue was measured 54 ± 8 and 20 ± 5 for the tissue around the vessel wall and 10 ± 1 for the lumen.

Chapter 7

Discussion & Conclusions

7.1 Discussion

The work presented in this thesis has included the basic principles behind the design and construction of transmit and receive arrays for imaging at ultra high fields. The focus was split between the electromagnetic profile of the probe and RF safety. The Larmor frequency increase at ultra high fields adds significant complexity to the design of RF coils.

The most common issue associated with UHF systems is the $\mathbf{B}_1$ field uniformity. Another equally prominent issue, one that arises at higher frequencies, is that human tissues become more power absorbent, i.e. exhibiting higher conductivities. Due to this fact, the re-calculation of the power limits, as they were set for standard clinical fields, is required. The construction of the device becomes more demanding too, since at 300 MHz for 7 T the coil becomes a more ‘coupled’ device; it is more prone to radiation loss, more sensitive to loading and external perturbation since the self resonance is closer to the Larmor frequency. In this thesis, all these issues have been detailed starting in chapter 4 for design and construction, in chapter 5 for patient safety and in chapter 6 which included a detailed characterisation as an RF and medical device.

In Chapter 4, the challenges of constructing such hardware were outlined. All recent and established concepts in coil building and RF design were employed in order to optimise both the transmit and receive chain. For the Tx array design, a reverse engineering methodology was followed starting from the particular application, i.e. black blood imaging. The RF pulse profile of the DANTE module was analysed and the required flip angles were translated to $\mathbf{B}_1^+$ field strength. Understanding the mechanism of blood suppression was necessary for also determining the extension of the RF field. In short, it was found that blood with a high velocity needs to be excited approximately 100 mm from the imaging plane (the bifurcations) towards the aortic arch. Finally, for the image acquisition a separate receive array was preferred in order to improve the sensitivity profile and also to benefit from parallel imaging.

In Chapter 5, RF safety was discussed in detail. The $\mathbf{B}_1^+$ inhomogeneity was addressed with multiple surface coils transmitting and generating a uniform field in the ROI. An RF shimming method has been utilised in this study as a tool to optimise the coil performance. The propagating waves from each channel can either cancel or add constructively creating local hot spots that can cause at the least discomfort if not skin burns.

In the safety chapter, all the necessary steps, as they have been presented and discussed by the UHF community, were taken to safeguard human subjects from excessive heating. A new methodology was also investigated, the virtual observation points (VOP), which, although promising, is still in its early stages to be trusted and employed and thus a conservative worst case scenario SAR supervision method was adopted instead. Another important step was the validation of the simulation results with experimental temperature measurements. The latter experiments were exhaustive and took place with independent coils as well as the constructed coil.

The worst case SAR supervision method was utilised with a power limit set per channel for 10 sec and 6 min averaging. The same limits were calculated for the global SAR but did not impose an additional penalty. Finally, healthy volunteers

were scanned for black blood and anatomical imaging. The results are deemed satisfactory and prove that the constructed coil can achieve the required $\mathbf{B}_1^+$ for blood suppression. Furthermore, suppression was achieved under a conservative power restriction regime, and the coil sensitivity profile showed potential for imaging with a higher resolution if the necessary optimisation steps are taken.

7.2 Future Work

The immediate intuition from looking at the presented results is whether the SNR can be further improved by optimising the sequence. The most straightforward method would be to determine carefully the imaging plane, to reduce the number of slices and to increase the number of averages in order to keep within a reasonable scan time. Different read outs can be used, more suited to 7 T as long as they can keep within the power restrictions.

Also, RF shimming for the presented results was made only for one slice in the imaging plane. It is known however that the two top coils contribute very little to the excitation before the read out and are there mainly to perform the blood nulling via the DANTE pulses. Thus, an RF shimming applied to the whole volume of the artery is desirable but due to the lack of receiving elements around that area this is definitely a challenging task. Also, $\mathbf{B}_0$ shimming, either static or dynamic, could be employed to reduce the susceptibility effects and T_2^* decay.

The initial intention for the receive array was for a 6 channel array with 3 channels on each side. The task proved challenging but since the cradle was designed to accommodate those coils they could be added on top of the four existing array elements. Also, the receive array tuning took place in the absence of the Tx array due to design restrictions. However, modifying the coil so that there is access to tune the receive loops in the presence of the de-tuned Tx side loops would be desirable and hopefully would improve the sensitivity.

Furthermore, the whole coil was designed with independently printed 3D pieces and each of them can be re-designed to fit another application or to improve a feature. For instance, the top compartment can be redesigned and constructed to fit another transmit loop extending the $\mathbf{B}_1^+$ even further down and thus allowing for more DANTE pulses and ideally better suppression. Additionally, the two side compartments could be made to slide in and out and lock at a certain distance from each other. That would make the coil suitable for a larger variation of neck length and diameter as well as improving patient comfort.

As far as safety is concerned, the coil was operated in a very conservative power supervision regime. Future work would certainly involve the validation and use of VOPs as the main supervision method. There is active research on how to experimentally certify the method and in collaboration with other labs this could be achieved. Ideally, for a number of random RF shims, the temperature increase measured on the hot spot in a phantom with known properties should match with the worst SAR report generated after each scan. The calculated SAR should also match the value reported from the scanner. If sufficient random shims follow these agreements then the VOPs could be deemed validated and used. That would allow a significant increase in power allowance per channel which would reduce scan time.

Finally, the coil SAR efficiency improvement scheme, calculated by $\mathbf{B}_1^+/\text{SAR}$, is another subject for a thorough investigation. The results are promising, including simulations and temperature measurements. Ideally, independent coils could be constructed each using a different shape, dimension and thickness for the plates as well as positioning over circular transmitting loops, always validated by simulations. Temperature measurements with and without the plates using phantoms with known properties on the hot spot can provide supporting evidence for the scheme. Finally, it could be possible to assess the validity of the hypothesis and the benefit gained from use in combination with VOP or other supervision schemes.

7.3 Conclusions

A 4Tx/4Rx coil was designed here to target the carotid arteries operating under pTx mode and a black blood imaging sequence was implemented for blood signal suppression and vessel wall imaging. The results from phantom and subject imaging show satisfactory flow and blood suppression for the whole cross sectional velocity profile. For the phantom measurements, the signal reduction from the circulating fluid while DANTE pulses remained was in the worst case 73.9% corresponding to a 40 cm/s velocity found in the diastolic phase in the carotid arteries.

In subject imaging, the high resolution readout used in combination with a DANTE module gave an average lumen SNR of 11 for both 2D and 3D sequences. The average tissue SNR was found to be 55 while the average vessel wall SNR was 24. The SNR for anatomical imaging was higher, averaging at 90, showing that the coil can be used for high resolution anatomical imaging. For 3 T DANTE black blood imaging the lumen SNR was 10, the static tissue SNR was 54, which are both similar values to 7 T. Also, 3 T images did not suffer from aliasing from parallel imaging as much as they did at 7 T but luckily, the ROI remained reasonably artefact free.

While the intrinsic magnetisation at 7 T is superior to that of 3 T, the performance as measured in both cases was similar. It could be argued that the suppression is more dependent on the coverage of the $\mathbf{B}_1^+$ regardless of the Larmor frequency and the intrinsic magnetisation. The full body transmit coil coverage in 3 T is allowing for very long DANTE pulse trains, ideally of low power, and hence affecting the static tissues less. The other advantage of 3 T at the moment is the possibility to run SE sequences since SAR can be effectively monitored. The restrictive power regime that surface Tx arrays operate at 7 T would never allow for a TSE excitation but, with future supervision regimes such as the VOPs, that might become available. Having comparable black blood imaging quality for the same parameters is nevertheless a step forward at 7 T showing the feasibility of the study and the

potential of the UHF strengths.

Finally, due to the use of a separate receive array the $\mathbf{E}$ and $\mathbf{B}$ fields of the Tx array were found to be reduced. However, it was also found that the coil demonstrated an improved $\mathbf{B}_1^+$ to SAR ratio compared to what it would have without the array present. On each side, the dual Rx loop in combination with the preamp ground planes in front of the transmitting capacitors reduced the $\mathbf{E}$ -field and hence reduced the SAR in proportion to the square of the $|E|$. A thorough simulation and experimental investigation has been made on the constructed coil. The results show a marginal SAR efficiency improvement when the worst case SAR is used and an overall minimum improvement of 11.4% when VOPs are used. The method could be generalised for the design and construction of more SAR efficient coils, i.e. ones able to achieve the same $\mathbf{B}_1^+$ for a lower SAR.

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