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Master of Science

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Abstract

Cell biological exposure studies in magnetic resonance imaging (MRI) environment, where a complex mixture of strong magnetic fields are present, have attracted considerable interest in recent years. The outcome of such studies might depend strongly on the conditions, for example exposure parameters and spatial variations of exposure. The aim of this thesis has been to give a detailed description of how the radio frequency (RF) magnetic field varies with position and sequence choice within an MRI bore from a patient perspective and to highlight the need of better consistency in future research.

Method: A straightforward theoretical description on the contribution to the RF magnetic field from a birdcage coil is given. A one dimensional coaxial loop antenna has been used as a probe to measure spatial variations of the RF magnetic field in a 1.5T MRI scanner. An exposure matrix containing RF magnetic field strength (\(H_1\)-field) amplitudes in three dimensions was constructed and used to study several clinical protocols and sequences. A qualified correspondence measurement was also made on a 3T MRI scanner.

Results: Around isocenter, for a common field-of-view (FOV), changes in exposure conditions were small; however, rapid changes of exposure conditions occurred upon approaching the end rings. The dominating \(H_1\)-field component switched from lying in the xy-plane to pointing the z-direction and was roughly 3 times larger than in isocenter. Practical difficulties indicate even larger differences at positions not measurable with the equipment at hand. The strongest \(H_1\)-field component was 32.6 A/m at position \((x,y,z)=(-24,8,24)\) cm from the isocenter.

Conclusions: Machine parameters such as repetition time, echo time and flip angle have little to do with actual exposure. Given specific absorption rate (SAR) values correlated well with the square of measured root-mean-square (RMS) values of the magnetic field (\(B_1^{2,RMS}\)) but not with peak values of the magnetic field (\(B_{1,peak}\)), indicating that peak values are not unlikely to be part of compromising factors in previous contradictory exposure research on genotoxicity. Furthermore exposure conditions depend strongly on position and unfavorable situations may occur in the periphery of the birdcage coil. Potentially elevated risks for conducting surfaces, for example arms or external fixations, in the proximity of the end rings, are proposed. Aside from spatial variation consideration on which type of geometry exposed cell-biological samples are placed in should be held since eddy currents, hot-spots and proper SAR depend on geometry. Conditions may vary considerably between in-vitro, ex-vivo and in-vivo studies since geometries of test tubes, petri dishes and humans differ.
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A Derivations

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Acronyms

BBM  Bessel Boundary Matching method. 15
BC  body coil. 7, 10, 11, 17, 26, 50
BCC  birdcage coil. 7, 10, 31, 50
CA  chromosomal aberrations. 1
CF cell  Crawford cell. 5, 6, 23, 24
CNS  central nervous system. 3
DC  duty cycle. 19, 20, 42, 47, 50
DSB  double strand breaks. 1
EM  electromagnetic. 5, 6
EMC  electromagnetic compatibility. 5
EMF  electromagnetic field. 1–3, 16, 17, 32, 50
EMP  effective measurement point. 31, 32
EPI  Echo Planar Imaging. 41, 42
FDFD  finite difference frequency-domain. 2
FDTD  finite difference time-domain. 2, 13, 15, 51
FEM  finite element method. 11, 13, 15, 51
FOV  field-of-view. iii, 34, 45
GF  gradient field. 1
GRE  Gradient Echo. 26, 31, 42, 47
MN  micro-nucleus. 1
MRI  magnetic resonance imaging. iii, 1–8, 11, 14, 17–19, 24–27, 30, 32, 40, 44–47, 50, 51, 53
OMP  original measurement point. 31, 32
PNS  peripheral nervous system. 1, 3
PtP  peak-to-peak. 5, 6, 42

RF  radio frequency. iii, 1–4, 7, 8, 11, 17–20, 23, 26, 27, 41–44, 47, 49, 51, 53

RMS  root-mean-square. iii, 3, 19, 41, 42, 53

SAR  specific absorption rate. iii, 1, 2, 18, 19, 40, 41, 50, 51, 53

SMF  static magnetic field. 1, 7, 26, 27

SNR  signal-to-noise ratio. 11

SSB  single strand breaks. 1

SSFP  Steady-state Free Precession. 27

TE  echo time. 3, 40

TEM  transverse electromagnetic. 5, 6

TM  transverse magnetic. 15

TR  repetition time. 1, 3, 40

TRUFI  True Fast Imaging with steady state free precession. 26, 27, 30, 31, 38, 41, 47

VNA  vector network analyzer. 49
1 Background

magnetic resonance imaging (MRI) is a fast emerging technique widely used in medical diagnostics all over the world. In comparison to other well-established medical imaging techniques like computed tomography, single positron emission computed tomography, positron emission tomography or conventional x-ray imaging which all incorporate ionizing radiation, MRI does not. In addition to this it is also considered a relatively safe, powerful and often indispensable diagnostic tool. Contrast and clinical status in MRI is obtained by applying a combination of three types of electromagnetic fields (EMFs), namely static magnetic fields (SMFs), gradient fields (GFs) and radio frequency (RF) fields, all of them classified as non-ionizing radiation. The biological effects of EMFs are known to some extent, for example: stimulation of the peripheral nervous system (PNS) and cardiac stimulation (Reilly, 1998) due to high slew-rate GF and thermal effects from RF magnetic fields (Šimunić, 2000). Some recent studies (Lee et al., 2011; Simić et al., 2008) suggest, however, that non-thermal, genotoxic effects may derive from clinical MRI exposure. The genotoxic effects observed include increased frequency of micro-nuclei (MN) following cell mitosis, chromosomal aberrations (CA), single strand breaks (SSB) (Simić et al., 2008) and double strand breaks (DSB) (Fiechter et al., 2013). MN frequency has further shown to correlate with cancer incidence according to a large cohort study by Bonassi et al. (2006). In contrast, a study by Szerencsi et al. (2013) suggests that DNA integrity of human leukocytes is not compromised following clinical MRI exposure. Genotoxic effects of SMFs alone have been given little attention (Feychting, 2005) but studies indicate that they are probably none or extremely small as reviewed by Ghodbane et al. (2013). There are clear indications that SMFs may alter cell growth and genotoxic effects caused by other factors such as ionizing radiation when they are used in combination. Similar effects regarding the combination of RFs and SMFs has not been shown. Therefore previously mentioned studies that have shown slight genotoxic effects following clinical MRI exposure of human lymphocytes (Lee et al., 2011; Simić et al., 2008) reveal GFs and RF magnetic fields as possible sources. Whether it is mainly one of these EMFs or a combination of the two which has shown an increase of MN, CA and SSB is thus far unknown.

The lack of characterization of EMF exposure parameters in clinical MRI environment is a deficit ascribed to present genotoxic studies. One could say that the preliminary code-of-practice in clinics is to vaguely categorize EMF exposure in the MRI by the number of scans, while in research by scan duration, repetition time (TR) and SAR for a set of clinical pulse sequences. Contradictory results from studies by Lee et al. (2011) and Szerencsi et al. (2013), the latter designed to partly reproduce the former by using similar pulse sequences (generated using different MRI scanners, General Electric HDx 3T versus Philips Achieva 3.0T), highlight the need of parametrized categorization of pulse sequences.
The World Health Organization (WHO) has recently identified needs of additional concern to different RF research topics (Saunders et al., 2010). Among other things WHO assigns high priority to dosimetric research assessing RF EMF emission characteristics, exposure scenarios and levels associated with body imaging. In order to obtain a dosimetric exposure model from a matrix complex of parameters following cell-biological as well as epidemiological cohort studies, the groundwork lies in parameterizing pulse sequences in terms of gradient magnetic field exposure and RF magnetic field exposure.

1.1 Previous work done in this field

To date, a literary review indicates that spatial exposure measurements of the RF magnetic field component have not been thoroughly done; however, work on numerical simulations relating to the subject is well examined (van den Bergen et al., 2009; Guler and Ider, 2012; Ibrahim et al., 2009, 2000, 2005). Yuan et al. (2012) constructed a tissue-mimicking phantom and compared an analytical solution of the heat transfer equation to temperature measurements with remarkable correspondence. This gives support to numerical simulations and highlight that heat dissipation is strongly connected to induced electric fields and SAR. A study by Kangarlu et al. (2007) compares induced electric field measurements with finite difference time-domain (FDTD) simulations in a phantom with considerable agreement. Simulations with FDTD and finite difference frequency-domain (FDFD) formulations are the main approaches in existing research. Such simulations depend on computer power and are time consuming. There are, in contrast to standard CPU based methods, GPU accelerated methods (Chi et al., 2010, 2011) for computation which show an $O(n^2)$ decrease in computation time. Numerical simulations have also been applied to examine the RF magnetic field and possible hot-spots around conducting metallic devices such as external fixations (Liu et al., 2013) and implants (Ballweg et al., 2011; Graf et al., 2006). Clearly availability, simplicity and flexibility of numerical simulations have made it the go-to technique in clinical patient-focused MRI safety research.

Some attempts to correlate $B_1$ FDTD simulations with $B_1$-mapping techniques, like the popular Bloch-Siegert shift method (Carinci et al., 2013; Sacolick et al., 2010) among others (Carinci et al., 2013; Homann et al., 2011; Katscher et al., 2009; Zhai et al., 2006), have also been made. In various situations this approach has been successful which is interesting since a possible correlation to the actual $B_1$-field could render it a directly employable method for spatial measurements of RF magnetic fields merely by using the scanner itself.
1.2 Aim

This Master’s thesis aims to increase understanding of RF magnetic field propagation and spatial variation inside an MRI (Siemens Magnetom Espree 1.5T, Siemens Healthcare, Erlangen, Germany) bore. Previous research relies significantly on simulations thus this work may be viewed as a “verification” measurement attempt similar to dose plan verification measurements in radiotherapy, for example (although this work is not performed on patient geometry). The idea is also for the result to strengthen basic guidance in future experiments, whether the RF magnetic field is examined or part of the source for cell-biological exposure and genotoxic studies. A risk is that studies performed today are not conducted under consistent conditions, even if believed so, as exposure parameters may depend heavily on the MRI scanner rather than parameters such as TR, echo time (TE), flip angle etc. Hence, even though equal sequences and protocols are used, the exposure can differ. The idea is furthermore to introduce isocentered measurements as a benchmark to identify maximum sequence exposure of the RF magnetic field, regardless of exposure maps being established by simulation or measurement.

Present knowledge tells that EMF interaction with tissue is described with completely different physics than ionizing radiation. The current dosimetric model used for EMF exposure relies on joule heating, dielectric heating, central nervous system (CNS) and PNS stimulation. None of these factors can, with current knowledge of the underlying physics, predict or describe results obtained by Fiechter et al. (2013); Lee et al. (2011); Simi et al. (2008), that is: knowledge of the complete interaction process is missing. For ionizing radiation dose calculation one does not actually need to measure anything since it may be acquired using Monte Carlo dosimetry simulations. It is indeed possible to simulate EMFs in various geometries and materials and hence also obtain exposure maps in e.g. the human body, but the elegant connection to dosimetry of possible carcinogenesis is not there yet. Based on the genotoxic studies already mentioned it should be said that: if a connection to carcinogenesis is found in future research, the link in between will probably be quite weak. Nonetheless a long time research goal may be to obtain similar models to that of ionizing radiation for predicting EMF exposure effects, but like many times before when it comes to physics, it is almost a need to know what one has: here meaning what kind of EMFs and how they are distributed. This thesis will focus on the $H_1$ (or $B_1$, RF, excitation) pulse and its spatial variation inside the MRI bore expressed in terms of its maximum amplitude. A relation to RMS values often employed elsewhere is simple to describe for fields with sinusoidal time dependence, as is the case here. The duty cycles for some different sequences in certain clinical protocols will also be presented in an exposure parameter table.

An inevitable question that has to be posed is: why not measure the conservative electric field instead of the RF magnetic field?. As proposed by Kangarlu et al. (2007), the induced electric field is the quantity
causing tissue heating and potential indirect tissue damage in MRI. Nonetheless it is suggested by Park et al. (2009) that the conservative electric field is at least an order of magnitude larger than the induced electric field. So the question seems adequate; however, the Siemens Magnetom Espree 1.5T MRI does incorporate an RF electric field shield (Figure 6). It should be left unsaid which MRI scanners that incorporate such a shield and which (if any) do not. The practical use in discussing the electric field here is therefore maybe not that interesting since such shielding has extremely high damping efficiency. The built-in RF shield in the MRI scanner used for this thesis is located on the inside of the gradient coil (see Figure 6). What remains is then the induced electric field originating from the RF magnetic field according to Faraday’s law of magnetic induction. This is why the RF magnetic field is measured.

1.3 Electric and magnetic field theory

The aim of this thesis is not to give a complete description of the theoretical magnetic field in MRI. It is however desirable to get some basic understanding on how to tie the results to coil construction. No other fields than the RF magnetic field or field strength will be considered. These are referred to as the $B_1$-field or $H_1$-field respectively and are interchangeable simply by the factor $\mu$. Sometimes $B_1$ will be used instead of $H_1$. Derivations of any non-intuitive formulas are found in Appendix A.

1.3.1 The magnetic field probe

The magnetic field probe is a Faraday shielded coupling loop of King-type model (Whiteside and King, 1964). It was constructed by Mikael Wendelsten of the radiophysics department at Umeå University, hence the name Wendelsten’s probe, and used previously by Sundström (2012) who gives it a more in-depth description. The idea is that a time varying magnetic field will induce a voltage difference between the sheath and inner conductor which can be measured with an oscilloscope. Considering Figure 9 a signal from the probe is obtained through Faraday’s law of induction, and is thus proportional to the rate of change, $\partial B / \partial t$, of the part of a magnetic field polarized in the direction in/out of the page. It is then convenient to ask how a measurement of this rate of change can be related to maximum magnetic field strength which, of course, can be totally independent of the rate of change. If one observes that the driving current, and hence also the magnetic field, varies sinusoidally it is clear that the calibration procedure will cover this: the time derivative in $\partial B / \partial t$ makes an $\omega$ pop out but information about $|B|$ remains. Obviously the same reasoning is not applicable to for example a square wave where the time derivative of the pulse theoretically will approach infinity regardless of its maximum amplitude. This is one of the reasons why the probe signal depends on $\omega$. Resonant modes of the probe may also influence signal strength. In general the probe dimensions, i.e. the diameter, has
to be small compared to the signal wavelength. This criteria is fulfilled here but it is still important to make a careful frequency dependent calibration to rule out eigenfrequency resonance as described in the next Section 1.3.2.

### 1.3.2 Calibration

The aim of calibrating the Wendelsten’s probe was to enable determination of the $H_1$-field distribution inside the MRI bore, not only raw measurement values (peak-to-peak (PtP) voltages). A transverse electromagnetic (TEM) cell, or Crawford cell (CF cell), was used for this task. A TEM cell is a device often employed in electromagnetic compatibility (EMC), i.e. immunity or emission, tests of electronic equipment. It generates electromagnetic (EM) fields of the TEM-mode and can thus approximate a far-field irradiation and plane wave propagation. Three radiation regions for antennas with its largest dimension, $d$, shorter than half the wavelength, $\lambda$, of the radiation it emits are defined as follows (Jackson, 1999):

- **Near (quasi − static) zone:** $d << r << \lambda$
- **Intermediate (induction) zone:** $d << r \sim \lambda$
- **Far (radiation) zone:** $d << \lambda << r$

where

\[
\lambda = \frac{c}{f},
\]  

(1.3.1)

$c$ is the speed of propagation and $f$ the frequency. Since the Larmor frequency (equation 1.3.8) in a 1.5T MRI will be around 63.8 MHz, a wavelength will correspond to roughly 4.7 meters. Evidently calibrations at these distances are impractical and probably even impossible since the source strength would have to be very large and reflections from the environment causing interference would compromise reliability. This is the reason why TEM cells are preferable. Figure 1 shows a sketch of a typical open CF cell like the one employed here. For more comprehensive material on CF cells the reader should look at the original technical note by Crawford and Workman (1979) and another thesis by Boriraksantikul (2008), the latter being very instructive and well written.
Since EM waves of the TEM mode are approximated, equations relating the Poynting vector $S$ to the electric and magnetic field strengths

$$ S = E \times H \quad (1.3.2) $$

become substantially simplified. The ratio of $E/H$ is then just the characteristic impedance of free space, $\sqrt{\frac{\mu_0}{\epsilon_0}} = 377 \Omega$, hence the above equation yields, for a far-field and plane wave approximation

$$ E = 377H. \quad (1.3.3) $$

In the near-field the coupling between $E$ and $H$ is much more complex since $E$ does not only oscillate in the transverse plane but in the direction of propagation as well. Therefore, reliable calibrations have to be made in the far-field. The electric field inside the CF cell is well defined and expressed by the ratio of the applied voltage, $V_{\text{in}}$, and the septum to outer plate separation distance, $l$

$$ E = \frac{V_{\text{in}}}{l} \quad (1.3.4) $$

A measurement in the CF cell cell yields a PtP voltage output, $V_{\text{PtP}}$, which is related to the $H$-field by a calibration factor, $k_{\text{cal}}$, with units $A/Vm$

$$ k_{\text{cal}} = \frac{V_{\text{in}}}{377 V_{\text{PtP}}} \quad (1.3.5) $$

This enables measurement of the magnetic field strength, $H_{1,MRI}$, without any knowledge of the nature of electric field strengths inside the MRI bore by the relation

$$ H_{1,MRI} = k_{\text{cal}} \cdot V_{MRI}. \quad (1.3.6) $$

This calibration factor applies to any relation between $H_1$ and $V$, regardless of maximum amplitude ($H_{1,\text{peak}}$) or root-mean-square ($H_{1,RMS}$) measurements.
1.3.3 Linear and circular polarization in a birdcage coil

The Espree Slim 70 BC incorporated in the MRI is of birdcage design and is schematically viewed in Figure 2.

Figure 2: Simple model of a birdcage coil (BCC). This particular coil has 16 rungs and 2 end rings. Such a BCC may either be a linearly polarized coil (single port excitation) or a circularly polarized coil (dual port, or quadrature, excitation). Tuning capacitors (not viewed in the figure) are placed along the rungs and rings to maintain the polarization as the Larmor frequency changes.

To obtain circular polarization two sinusoidal driving currents with a 90° phase shift placed orthogonal to each other is used (generally). The requirement for the sinusoidal driving currents to be maintained is that the total added phase shift (not to be mistaken for the 90° phase shift between the driving currents) along N equally spaced rungs make up an integer multiple of 2π for a complete lap around the end rings

\[ N \Delta \phi(\omega) = 2\pi M. \]  

(1.3.7)

The most homogeneous \( B_1 \) field is produced in the \( M = 1 \) mode, where an alternative way of expressing this is that the signal wavelength equals the end ring circumference, taking capacitors into account and assuming equal rung spacing. Another requirement for image acquisition is that the driving current frequency, dictating the RF field frequency, equals the Larmor frequency \( \omega \). Since the slice selection gradient, \( G_z \), slightly shifts the Larmor frequency along the z-axis the BCC needs to adjust its resonant frequency as well. This is achieved by using tuning capacitors. The relation to the Larmor frequency is

\[ \omega = \gamma (B_0 + \Delta z G_z). \]  

(1.3.8)

where \( \gamma \) is the gyro-magnetic constant, \( B_0 \) the SMF (approximately 1.5T) and \( \Delta z \) the displacement from \( z = 0 \).

The terms linear and circular polarization are well known to most people with physics background; however, their interpretation in the MRI and particularly the BCC is non-intuitive. It is convenient to separate the lab and the rotating frame of reference. The rotating frame of reference is where the magnetization vector at Larmor precession around \( B_0 \) exists. To flip this magnetization into the
xy-plane the $B_1$ component that will dictate the flip angle must also rotate with the same angular frequency and in the same direction as the magnetization vector. It is then convenient to ask oneself: how is this accomplished with linear polarization? The answer is in fact simple in this context. For linear polarization the wave can be decomposed into two counter-rotating components generally referred to as the $B_1^+$ and $B_1^-$ component. This holds for each and every field produced from the current in the rungs. In a tuned (at resonance) linearly polarized coil, which is fed at one port, a standing wave along the coil circumference is produced. Consider the feeding current density $J_{\text{port}}(t) = J_{\text{port}} e^{i(\omega t - \delta)} \hat{z}$

$$J_{\text{port}} = J_{\text{port}} e^{i(\omega t - \delta)} \hat{z}$$

(1.3.9)

This means that depending on where the feeding port is, the RF field with be polarized perpendicular to an imaginary line between the feeding port and the coil center, following the fact that some rungs will always give zero contribution to $B_1$ (as will be shown momentarily). Notice also that the $r$ dependence of $J$ has been dropped, following an assumption of uniform current distribution, and because the MRI bore geometry can be expressed with the aid of another function $f(n)$; however, it should be said that the position vector of rung $n$ is $R_n = R [\cos ((n - 1)\Delta \phi(\omega))\hat{x} + \sin ((n - 1)\Delta \phi(\omega))\hat{y}]$ implying that rung 1 is located in the direction of $\hat{x}$ and rung 5 in the direction of $\hat{y}$ and so on (refer to Figure 4). The real contribution to $B_1$ from rung $n$ is proportional to the current in rung $n$ which is given by

$$J_{\text{port}}^n = J_{\text{port}} f(n) e^{i(\omega t - \delta)} \hat{z}.$$  

(1.3.10)

For the $0^\circ$-port $J_{n}^{0^\circ} = \cos ((n - 1)\Delta \phi(\omega))$ and $\delta = 0$, thus

$$\Re\{J_{n}^{0^\circ}(t)\} = \Re\{J_{n}^{0^\circ} \cos ((n - 1)\Delta \phi(\omega)) e^{i(\omega t)}\} \hat{z}$$

$$= J_{n}^{0^\circ} \frac{1}{2} [\cos (\omega t - (n - 1)\Delta \phi(\omega)) + \cos (\omega t + (n - 1)\Delta \phi(\omega))] \hat{z}$$

$$= J_{n,-}^{0^\circ}(t) + J_{n,+}^{0^\circ}(t)$$

(1.3.11)

This contribution is generally present for both linear and circularly polarized birdcage coils. These currents generate what was earlier described as the co-rotating, $B_1^+$, and counter-rotating, $B_1^-$, part of the $B_1$-field. An example of how this current and its decompositions look like around the coil circumference for one period can be seen in Figure 3. The technique is called single port linear excitation.

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1 This is not entirely true: the current density is also modulated differently depending on which sequences and which parameters are chosen. This is an amplitude modulation with time dependence $(C(t))$ which would propagate into the solutions to 1.3.33 and 1.3.34 and particularly render different expressions in the time derivatives of $J$. For the purpose of this thesis this is unnecessary to consider.
In a circularly polarized coil, as in the case here, two feeding ports with a 90° current phase shift placed orthogonal to each other are used (there exists in fact birdcage coils with a single feeding port which also produce circular polarization). Figure 4 gives a brief view of how the rungs are located.

This technique is called quadrature excitation and is common knowledge in radar transceiver electromagnetics. The driving current feed to the orthogonal driving port, \( J_n^{90\circ}(t) \), is also given by equation 1.3.9 but now with \( \delta = \pi/2 \) and \( f(n) = \cos((n+3)\Delta\phi(\omega)) \). Hence the contribution to rung \( n \) from
this orthogonal driving port is:

\[ J_{n}^{90^\circ}(t) = J^{90^\circ}\cos((n+3)\Delta\phi(\omega))e^{i(\omega t-\pi/2)}\hat{z} \]  \hspace{1cm} (1.3.12)

In co-action with equation 1.3.11 the current feed to a rung located orthogonal to rung \( n \), expressed by equation 1.3.13, is what produces the circularly polarized field. Hence, for an \( N=16 \) BCC (which is the case for the Espree Slim 70 BC), rung \( n = 5 \) will be located orthogonal to rung \( n = 1 \). Note that knowledge of the exact location of the feeding ports are not claimed, but merely their separation of \( 90^\circ \) on a supposable ring. The real contribution becomes

\[ \Re\{J_{n}^{90^\circ}(t)\} = \Re\{J^{90^\circ}\cos((n+3)\Delta\phi(\omega))e^{i(\omega t-\pi/2)}\} \hat{z} = \]

\[ J^{90^\circ}\frac{1}{2}[\cos(\omega t-\pi/2-(n+3)\Delta\phi(\omega)) + \cos(\omega t-\pi/2+(n+3)\Delta\phi(\omega))] \hat{z} = J_{n,-}^{90^\circ}(t) + J_{n,+}^{90^\circ}(t) \]  \hspace{1cm} (1.3.13)

A schematic view of how the standing waves from Figure 3 in linearly excitation combine to yield circular polarization in quadrature excitation can be seen in Figure 5.

Figure 5: An illustration of quadrature excitation. Standing waves from the \( 0^\circ \)-port (solid line) and \( 90^\circ \)-port (dashed line) are added together by the superposition principle. The orthogonal \( 90^\circ \)-port is phase shifted \( 90^\circ \) ahead of the other, making the resultant wave travel to the left through time instances 1-16. The ports are marked as black dots.
To sum up one finds the total real current in quadrature excitation around the coil

\[ J_n(t) = \Re\{J_n^0(t) + J_n^{90\circ}(t)\} = J_{n,+}(t) + J_{n,-}(t) + J_{n,+}^{90\circ}(t) + J_{n,-}^{90\circ}(t) \] (1.3.14)

which should give the reader a good understanding on why circularly polarized coils are more power efficient than linearly polarized coils (signal-to-noise ratio (SNR) improved by a factor \(\sqrt{2}\)), and hence also able to reduce the RF magnetic field exposure to the patient. In words only the co-rotating part is useful for MRI, since this is the part that can excite the proton spins\(^2\). In linear excitation we have two parts where the co-rotating part only has half the amplitude of the standing waves, which in quadrature excitation are used to their maximum since there is no counter-rotating part. This is why the peak power input can be lowered by approximately a factor \(1/\sqrt{2}\) for quadrature coils compared to single port linear coils whilst maintaining roughly the same SNR. The reader is referred to some neat COMSOL Multiphysics simulations made by Gurler and Ider (2012) on this matter. They present several instructive plots which compare the field strengths for a simple set-up of linear and quadrature excitation respectively.

### 1.3.4 Magnetostatic approximation of the \(B_1\)-field in MRI

To give the reader some basic understanding on why the results in Section 3 look like they do, without rushing through tedious mathematical physics, the magnetostatic case of Ampere’s law, namely the Biot-Savart law for a contribution to \(B_1\), may be considered

\[ B_1(r) = \frac{\mu_0}{4\pi} \int_C \frac{dl \times \mathbf{s}}{r^3} \] (1.3.15)

where \(dl\) is an infinitesimal curve segment and \(I\) the current, assuming uniform current density distribution across a conducting cross-section of the rung and end rings. To a point \(r\) in space the contribution to \(B_1\) will then be governed by the birdcage design. The shady symbol \(\mathbf{s}\) is called the separation vector and is defined as the vector pointing from the source to the field point (Griffiths, 1999).

\[ \mathbf{s} \equiv r - R \] (1.3.16)

where \(r\) points to the field point and \(R\) to the source. The Biot-Savart law is analytically solvable only for simplistic geometries, otherwise one has to employ numerical methods such as finite element methods (FEMs). Such a simulation is not considered in this thesis; however, it has been done multiple times before in the time dependent case as mentioned in 1.1. To get understanding of the \(B_1\) magnitude.

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\(^2\) This is a simplification. For image acquisition the BC shifts between transmit and receive mode. For quadrature coils this is accomplished by phase shifting the driving current to for example the orthogonal 90\(^\circ\)-port by -180\(^\circ\) so that it is excited with a phase shift of -90\(^\circ\) relative to the 0\(^\circ\)-port. This has no effect on \(B_1\) magnitude.
distribution some simple cases of $\mathbf{B}$-fields around straight wires and circular loops may be considered.

For the rungs, the sum over $N$ straight wire segments breaks down to

$$\mathbf{B}_{1}^{\text{rung}}(\mathbf{r}) = \sum_{n=1}^{N} \mathbf{B}_{1,n}(\mathbf{r}) = \sum_{n=1}^{N} \frac{\mu_0 I_n}{4\pi s_n} (\sin \theta_{2,n} - \sin \theta_{1,n}) \hat{z} \times \hat{s}_n$$  \hspace{1cm} (1.3.17)

where $I_n$ is expressed by the static versions of equations 1.3.11 and 1.3.13, $\theta_{2,n}$ is the angle to the end of, and $\theta_{1,n}$ the angle to the start of, the wire segment (as seen from a point at a right angle distance $s$ from wire $n$). A more comprehensive way of expressing this is to start with the vector potential from a segment of straight wire

$$\mathbf{A}_{1,n}(\mathbf{r}) = \frac{\mu_0 I_n}{4\pi} \int_{z_1}^{z_2} \frac{\hat{z}}{s_n} dz'$$  \hspace{1cm} (1.3.18)

where $z_1$ and $z_2$ are distances to the start- and endpoints of each respective wire segment. $\mathbf{B}_{1,n}$ may then be found by

$$\mathbf{B}_{1,n}(\mathbf{r}) = \nabla \times \mathbf{A}_{1,n}(\mathbf{r}).$$  \hspace{1cm} (1.3.19)

The answer, after some algebra, is

$$\mathbf{B}_{1}^{\text{rung}}(\mathbf{r}) = \sum_{n=1}^{N} \mathbf{B}_{1,n}(\mathbf{r}) = \sum_{n=1}^{N} \frac{\mu_0 I_n}{4\pi (x^2 + y^2 + R^2 - 2Rx \cos (n\Delta \phi) - 2Ry \sin (n\Delta \phi))} \int_{z_1}^{z_2} \frac{\hat{z}}{s_n} dz' \cdots \frac{L - z}{(z^2 + x^2 + y^2 + R^2 - 2Rx \cos (n\Delta \phi) - 2Ry \sin (n\Delta \phi))^{1/2}} + \cdots \frac{z}{(z^2 + x^2 + y^2 + R^2 - 2Rx \cos (n\Delta \phi) - 2Ry \sin (n\Delta \phi))^{1/2}} \cdots \left[-(y - R \sin (n\Delta \phi)) \hat{x} + (x - R \cos (n\Delta \phi)) \hat{y} \right]$$  \hspace{1cm} (1.3.20)

which may look messy but contains all necessary variables needed to calculate the contribution to $\mathbf{B}_1$ from the rungs at the point $\mathbf{r} = (x, y, z)$ in the magnetostatic case. Here $R$ is the coil radius, $L$ the rung length and $z = 0$ defined as the plane where one of the end rings is located. One may also note that $\mathbf{R}_n = R \cos (n\Delta \phi) \hat{x} + R \sin (n\Delta \phi) \hat{y} + z' \hat{z}$.

In the same manner, for the end rings, the Biot-Savart law for a circular loop expressed component wise yields

$$B_{1,x}^{\text{ring}} = \frac{\mu_0 I}{4\pi} Rz \int_{0}^{2\pi} \frac{\cos \phi d\phi}{(x^2 + y^2 + z^2 + R^2 - 2Rx \cos \phi - 2Ry \sin \phi)^{3/2}}$$  \hspace{1cm} (1.3.21)

$$B_{1,y}^{\text{ring}} = \frac{\mu_0 I}{4\pi} Rz \int_{0}^{2\pi} \frac{\sin \phi d\phi}{(x^2 + y^2 + z^2 + R^2 - 2Rx \cos \phi - 2Ry \sin \phi)^{3/2}}$$  \hspace{1cm} (1.3.22)

$$B_{1,z}^{\text{ring}} = \frac{\mu_0 I}{4\pi} R \int_{0}^{2\pi} \frac{R - x \cos \phi - y \sin \phi d\phi}{(x^2 + y^2 + z^2 + R^2 - 2Rx \cos \phi - 2Ry \sin \phi)^{3/2}}$$  \hspace{1cm} (1.3.23)

where $R$ is again the coil radius and $(x, y, z)$ the point position in space. These are elliptic integrals and lack solutions that can be expressed solely with elementary functions; however, a little aid for
better understanding can be obtained by noticing the fact that one may choose a coordinate system which places the point at say \((0,y,z)\). This makes all factors containing \(x\) vanish, and also renders the expression for \(B_{1,x}^{\text{ring}}\) to be solved. The answer is the trivial solution \(B_{1,x}^{\text{ring}} = 0\). A coordinate system specific to any point may be chosen in this way, and one may note that only a radial and \(z\)-component comes from the end rings, that is: no component in the direction of the instantaneous axis of rotation. To sum up we have: radial and axial contributions from the rungs plus radial and \(z\)-axis contributions from the end rings in the magnetostatic approximation. The reader should now have obtained sufficient knowledge to understand the results. In addition the complete solution to a harmonically oscillating source is fortunately within reach for a simplified geometry. For other non-simplified cases of geometry FDTD and FEM with different theoretical starting points are generally applied.

### 1.3.5 General solution for electromagnetic radiation in MRI

In the static case the Biot-Savart law provided a solution to \(B_1(r)\) for constant currents. In the time-dependent configuration, electrodynamics; however, the answer is a bit more tedious following several different observations: causality is now a factor, the \(E(r,t)\) and \(B(r,t)\) components are considered duals thus affecting each other and \(E\) can no longer be expressed as the gradient of a scalar potential, i.e. \(E = -\nabla V\), since the curl of \(E\) is non-zero. Hence the total description of the fields are governed by the general solution to Maxwell’s equations. The solution can be obtained in different ways, normally by, but not limited to, using potential formulation or by solving Jefimenko’s equations for the fields directly where the former indirectly generates the latter. Let’s start by remembering Maxwell’s equations (the 1 index is dropped since no other fields than \(B_1\) are considered)

\[
\begin{align*}
(i) \quad & \nabla \cdot E = \frac{\rho}{\epsilon_0}, \quad (iii) \quad \nabla \times E = -\frac{\partial B}{\partial t}, \\
(ii) \quad & \nabla \cdot B = 0, \quad (iv) \quad \nabla \times B = \mu_0 J + \mu_0 \epsilon_0 \frac{\partial E}{\partial t}.
\end{align*}
\]

(1.3.24)

One first notes that \(B\) has zero divergence, just as in the magnetostatic approximation, so equation 1.3.19 is still valid. Faradays’ law (equation 1.3.24 (iii)) then yields

\[
\nabla \times \left( E + \frac{\partial A}{\partial t} \right) = 0.
\]

(1.3.25)

By noting that the curl of the gradient of a scalar potential, \(\phi\), is zero it is possible to choose

\[
E = -\nabla \phi - \frac{\partial A}{\partial t}
\]

(1.3.26)
which may then be substituted into Gauss’s law (equation 1.3.24 (i))

\[
\nabla^2 \phi + \frac{\partial}{\partial t} (\nabla \cdot \mathbf{A}) = -\frac{\rho}{\epsilon_0}.
\]

(1.3.27)

By using the chosen \( B \) and \( E \) expressions from equations 1.3.19 and 1.3.26 the Ampère/Maxwell law (equation 1.3.24 (iv)) may be written as

\[
\left( \nabla^2 \mathbf{A} - \frac{1}{c^2} \frac{\partial^2 \mathbf{A}}{\partial t^2} \right) - \nabla \left( \nabla \cdot \mathbf{A} + \frac{1}{c^2} \frac{\partial \phi}{\partial t} \right) = -\mu_0 \mathbf{J}
\]

(1.3.28)

where \( c = \sqrt{\mu_0 \epsilon_0} \). Not only do these two last equations reduce the initial number of six problems down to four, but they can be manipulated further to obtain neat expressions. One may exploit the advantages of \textit{gauge freedom}, particularly by choosing the \textbf{Lorentz gauge}

\[
\nabla \cdot \mathbf{A} = -\frac{1}{c^2} \frac{\partial \phi}{\partial t}.
\]

(1.3.29)

This means that the scalar potential and the vector potential will be preceded by the same differential operator called the \textbf{d’Alambertian}

\[
\Box^2 \equiv \nabla^2 - \frac{1}{c^2} \frac{\partial^2}{\partial t^2}
\]

(1.3.30)

reducing all information contained in the Maxwell’s equations down to

\[
\left\{\begin{array}{l}
(\text{i}) \quad \Box^2 \phi = -\frac{\rho}{\epsilon_0} \\
(\text{ii}) \quad \Box^2 \mathbf{A} = -\mu_0 \mathbf{J}
\end{array}\right.
\]

(1.3.31)

which are two inhomogeneous wave equations. As a little side note, these two equations may be reduced to one by considering the four-vector potential, \( A^\mu = (V/c, A_x, A_y, A_z) \), especially practical in the relativistic approach. The relativistic solution is; however, unnecessarily ambitious to derive within the scope of this thesis since, well, a patient lies considerably still during an MRI scan. Anyhow it is from a theoretical point of view important to incorporate the causality factor into the solutions to equations 1.3.31 (i) and (ii). This is done by introducing the \textbf{retarded time} \( t_r \equiv t - \hat{r}/c \) and the \textbf{retarded potentials}

\[
\phi(\mathbf{r}, t) = \frac{1}{4\pi\epsilon_0} \int \frac{\rho(\mathbf{r}', t_r)}{\hat{r}} \, dV', \quad \mathbf{A}(\mathbf{r}, t) = \frac{\mu_0}{4\pi} \int \frac{\mathbf{J}(\mathbf{r}', t_r)}{\hat{r}} \, dV'.
\]

(1.3.32)
which give adequate solutions for \( \mathbf{B} \) and \( \mathbf{E} \) when applying equations 1.3.19 and 1.3.26. The results are the Jefimenko’s equations:

\[
\mathbf{B}(\mathbf{r}, t) = \frac{1}{4\pi\varepsilon_0} \int \left[ \frac{\mathbf{J}(\mathbf{r}', t)}{r^2} + \frac{\dot{\mathbf{J}}(\mathbf{r}', t)}{cr} \right] \times \mathbf{\hat{r}} \, dV' \tag{1.3.33}
\]

and

\[
\mathbf{E}(\mathbf{r}, t) = \frac{\mu_0}{4\pi} \int \left[ \frac{\rho(\mathbf{r}', t)}{r^2} + \frac{\dot{\rho}(\mathbf{r}', t)}{cr} - \frac{\dot{\mathbf{J}}(\mathbf{r}', t)}{c^2 r} \right] \times \mathbf{\hat{r}} \, dV' \tag{1.3.34}
\]

which may be solved by using the expression for the total current of the rungs and rings respectively, in the same manner as in the magnetostatic approximation. As to what the Biot-Savart law tells us, the resemblance between the static and quasi-static approximation is governed merely by a time dependence. The Jefimenko’s equations will not be solved here because it is simply not relevant for the outcome of this thesis. It is not relevant because the Jefimenko’s equations by themselves yield such strong support for the quasi-static theory that the magnetostatic approximation and its resemblance to the quasi-static case is enough. The reader might ask why this is. The magnetic field case of Jefimenko’s equations may be considered. If one were to Taylor expand the \( \mathbf{J}(\mathbf{r}', t) \)-term to first order, hence dropping any higher order terms

\[
\mathbf{J}(\mathbf{r}', t_r) = \mathbf{J}(\mathbf{r}', t) + (t_r - t)\dot{\mathbf{J}}(\mathbf{r}', t) + ...
\]

a fortuitous cancellation between the second term of equation 1.3.33 (to first order), \( \dot{\mathbf{J}}(\mathbf{r}', t)/cr \), and \((t_r - t)\dot{\mathbf{J}}(\mathbf{r}', t)/r^2 \) occurs and one is left with the quasi-static version of the Biot-Savart law. Evidently, more exact expressions than those already given are excessive to understand the results.

### 1.3.6 Additional notes on numerical solutions

Numerical methods, such as FDTD, FEM or the Bessel Boundary Matching method (BBM), rely on different techniques to solve Maxwell’s equations. FDTD techniques are based upon relationships between \( \mathbf{B} \) and \( \mathbf{E} \) field components and updated with iterative methods. The BBM methods uses a different approach, with Maxwell’s equations solved for the transverse magnetic (TM) mode and expressed by Bessel/Fourier series. These methods takes their starting points in equation 1.3.31. If one considers the cylindrical coordinate system, so that \( \mathbf{A} = \mathbf{A}(r, \theta, z, t) \) and \( \phi = \phi(r, \theta, z, t) \), and notes that the free current is \( \mathbf{J}(\mathbf{r}, t) = \mathbf{J}(\mathbf{r}, t)_{\text{ind}} + \mathbf{J}(\mathbf{r}, t)_{\text{ext}} \) (the sum of induced and external current), Maxwell’s equations can be used to describe the fields in a variety of matter. For air, as is the case in all measurements in this thesis, the electrical conductivity, \( \sigma \), is practically zero so the induced current may be neglected; however, for a more thorough interpretation it is kept, also: each rung (and
end ring) actually effects the other rungs or rings by inducing currents in them yielding secondary, third, fourth etc. party contributions (although probably very small). In work presented by van den Bergen et al. (2009) the gauge freedom is used to eliminate the scalar potential, leaving only the vector potential \( A \) to describe the fields. The equation is a Helmholtz (or eigenvalue) equation, where \( J(r, t) \) is the free current as described above

\[
\epsilon \mu \frac{\partial^2 A}{\partial t^2}(r, t) = \nabla^2 A(r, t) + \mu J(r, t). \tag{1.3.36}
\]

Since the scalar potential has been eliminated by the gauge freedom equation 1.3.26 may be used to express the induced current

\[
J(r, t)_{\text{ind}} = \sigma E = -\sigma \frac{\partial A}{\partial t}(r, t) = -i\omega \sigma A(r, t) \tag{1.3.37}
\]

One may extract the information that an \( i\omega \) pops out from the time derivative when using complex notation of the fields which enable further simplifications by introducing \( \xi^2 = \epsilon \mu \omega^2 - i\omega \sigma \mu \) to equation 1.3.36 after plugging in the expression for \( J(r, t)_{\text{ind}} \). The expression is

\[
\xi^2 A + \nabla^2 A = -\mu J_{\text{ext}} \tag{1.3.38}
\]

which is discussed in more detail by van den Bergen et al. (2009). Some ideas on how rung and ring currents may be expressed are

\[
J(r, t)_{\text{rungs}} = \sum_{n=1}^{N} I_n e^{i(\omega t + \phi)} \delta(r - R_n) \delta(\phi - \phi_n) \hat{z} \tag{1.3.39}
\]

and

\[
J(r, t)_{\text{rings}} = \sum_{k=1}^{K} I_k e^{i(\omega t + \phi)} \delta(r - R) \delta(z - z_k) \hat{\phi}. \tag{1.3.40}
\]

The general solutions are then expressed with the aid of first and second kind Bessel functions in different regions (air or tissue regions that may or may not include \( J(r, t)_{\text{ext}} \)) which enable improved computation speed compared to direct numerical solutions of Jefimenko’s equations. As usual only the real part has physical meaning.

### 1.3.7 Biological extension

Time varying EMFs induce electric fields and electric currents in the body and interact with tissue according to three coupling mechanisms (International Commission on Non-Ionizing Radiation Protection (ICNIRP), 1998).
Low frequency (LF) electric fields interaction with tissue implies generation of electric current, bound charge polarization and reorientation of electric dipoles.

LF magnetic fields are coupled to the induction of electric fields in the human body.

Energy absorption from EMFs is significant when frequencies above 100 kHz are used, like in the MRI environment. The human body is complex not only in the sense of possessing both conductive and dielectric properties, but also because innovative orthopedic and surgical procedures introduce additional non-body (however often body-like) material. The induction of electric fields originating from external RF magnetic fields in tissue may from a simplistic point of view be described by remembering Faraday’s law of induction over a closed loop $\partial \Sigma$ encompassing the surface $\Sigma$

$$\oint_{\partial \Sigma} \vec{E} \cdot d\vec{l} = -\frac{d}{dt} \iint_{\Sigma} \vec{B} \cdot d\vec{A} = -\frac{d\Phi_B}{dt}$$

resulting in

$$\tilde{E}2\pi r = \frac{d\tilde{B}}{dt} \pi r^2 = -i\omega B \pi r^2.$$  \hspace{1cm} (1.3.42)

Taking the real part, given $\tilde{E}$ and $\tilde{B}$ in complex notation, one finds

$$J = \sigma \Re[\tilde{E}] = \Re[-i\omega \tilde{B}] \frac{\sigma r}{2} = \Im[\tilde{B}] \sigma \pi f r$$ \hspace{1cm} (1.3.43)

as the induced current density. Actually this is not entirely true: since the material of interest is conducting, skin depth should be considered. This means that one would have a radial dependence of induced eddy currents. This is described in-depth by Bottomley and Andrew (1978) with the solution

$$J = \sigma \Re[\tilde{E}] = \Re[-i\omega \tilde{B}] \sigma \frac{I_1(\xi r)}{I_0(\xi r_0)} \hat{\theta} = \Im[\tilde{B}] \sigma \omega \frac{I_1(\xi r)}{I_0(\xi r_0)} \hat{\theta}$$ \hspace{1cm} (1.3.44)

where $K$ is the complex wavenumber, $I_0$ and $I_1$ the Bessel functions of the first kind of order zero and one with independent variables $\xi r$ and $\xi r_0$ respectively. Refer to Jackson (1999) for a description of Bessel functions as a solution to the modified Bessel equation. Either way, the magnetic flux density may be related to the magnetic field strength by the magnetic permeability as

$$B = \mu H$$ \hspace{1cm} (1.3.45)

where $H$ is the quantity calibrated for in this thesis as explained in Section 1.3.2. The electric field component also affects the human body; however, the Espree Slim 70 BC has an E-field shield (Figure 6) plus it is outside the scope of this thesis.
1.3.8 Thermal effects

Thermal effects are not considered in the measurements but it might be useful for the reader to get a tool spanning the concept of tissue heating and measured fields. Regarding RF electric and magnetic field exposure from medical equipment, a quantity currently used to define exposure limits is the SAR-value (International Electrotechnical Commission (IEC), 2010). Mitigation is primarily directed towards tissue heating since the SAR value is strongly linked to tissue temperature effects. The SAR value expresses the absorbed power per mass and is defined as

\[
SAR = \frac{d}{dt} \left( \frac{1}{\rho} \frac{dW}{dV} \right)
\]  

(1.3.46)

where \(dW\) is the energy absorbed in the volume element \(dV\) of density \(\rho\). Calculation of the SAR value is possible if the electric fields within the volume is known

\[
SAR = \int_{\text{sample}} \frac{\sigma(r) |E(r)|^2}{\rho(r)} dr
\]  

(1.3.47)

In most cases there are needs to distinguish between two different sources of tissue heating, namely dielectric and induction heating. The model for describing the generated power density per unit volume for dielectric heating is vaguely explored. It has a complex relation to conductivity and permittivity of the medium and frequency of the electric field. Small conductivity or high frequency are indications that dielectric heating is the dominating factor regarding energy loss to the medium. For conducting media, like metal implants, induction heating is dominant. Here the resistive losses of the produced eddy currents show as thermal energy. In the MRI environment interaction between RF electric or magnetic field and tissue occurs in the near-field region, where SAR frequency dependence
curves as presented by the International Commission on Non-Ionizing Radiation Protection (ICNIRP) (1998) are no longer valid. Considering near-field interaction the suggestion is to either continue using whole-body SAR restrictions (a member to basic restrictions) or to use presented field exposure levels (a member to reference levels) for the $E$ and $H$ fields independently since their contributions of deposited energy cannot exceed the SAR restrictions. The current standards for assessing patient safety in the vicinity of electric medical devices are the IEC international standards (International Electrotechnical Commission (IEC), 2010). The exposure limits defined by the IEC is a syndicate of research, including the ICNIRP guidelines, dedicated to patient and worker safety that apply to MRI. No in-depth information from these standards will be presented here since it is enclosed by strict copyright regulations; however, some basics limits are given in Table 1 as a reference to the results.

Table 1: Whole body SAR limits in current practice

<table>
<thead>
<tr>
<th>OPERATING MODE</th>
<th>6 MIN AVERAGE SAR LIMIT</th>
<th>SHORT DURATION (10s) SAR LIMIT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal</td>
<td>2 W/kg</td>
<td>4 W/kg</td>
</tr>
<tr>
<td>First level controlled</td>
<td>4 W/kg</td>
<td>8 W/kg</td>
</tr>
<tr>
<td>Second level controlled</td>
<td>&gt;4 W/kg</td>
<td>&gt;8 W/kg</td>
</tr>
</tbody>
</table>

There are also body temperature rise regulations which are assumed to be met by handling the SAR limits. Higher level operating modes are used for research purposes or in examinations believed to give clinical information sufficiently valuable to motivate a risk of possible damage caused by increased exposure. Exposure limits are generally given as a 6 minute average or this 6 minute average limit times 2 for any 10 second period. Also there are, as mentioned earlier, unsatisfactory documentation on the effects of local tissue heating and local tissue exposure overall, even though local elevations in SAR levels observed in anatomical human models (Murbach et al., 2011; Nadobny et al., 2007; Neufeld et al., 2011) are evident. These are the primary reasons why criticism to whole-body SAR restrictions for application in MRI safety is raised (van Rhoon et al., 2013). RF pulses in MRI may have duty cycles (DCs) of as low as 1% which combined with very high magnetic field peaks still may render the SAR and $B_{1,RMS}$ limits to be met. The whole-body SAR will nonetheless be presented as a parameter in this thesis since it is currently the standard approach to assess patient safety according to the International Electrotechnical Commission (IEC) (2010); however, skepticism to the SAR parameter should be kept in mind. Some suggestions have been made to assess part of the inadequacy of SAR restrictions in MRI, one of them being the CEM43°C thermal dose threshold discussed by van Rhoon et al. (2013). The International Electrotechnical Commission (IEC) (2010) also highlights that their standards might be updated in the near future with the CEM43°C model taken into consideration.

The RMS value may give a measure of average exposure over time, but might also be insufficient to explain results from previous studies discussed in Section 1. Another parameter of interest, aside from

---

3 Note specifically that the ICNIRP guidelines apply to the general public and not MRI, where the IEC international standards apply instead. Both are partly based on common research research.
peak values, is the DC. One may define DCs in different ways, where the most common definition is the fraction of the total time the pulse is active. This definition is not used because it is most applicable to square signals. Instead, another definition is introduced: the real duty cycle, $DC_r$. It has been used earlier in a related Master’s thesis by Sundström (2012) but is here refined to being the actual real DC rather than a triangular approximation

$$DC_r \equiv \pi V_{peak}^{-1} \int_{t_0}^{t_N} |V(t')| dt'.$$

(1.3.48)

Shapes of the RF pulses differ considerably (see Figures 7 and 8) depending on which sequence is chosen, hence they cannot be approximated in a general fashion. The digital oscilloscope employed does however provide sufficient information to make numerical calculations on the actual shapes. The $DC_r$ for a pulse can be approximated with a one dimensional measurement simply by realizing that even though one gets a sinusoidal readout on the oscilloscope, the pulse is fully occupied during its duration due to circular polarization. This is why the factor $\pi$ comes into play (the integral of $|\cos(x)|/\int_0^{2\pi} 1 d\xi$ over one period is $2/\pi$ and the amplitude taken as $V_{peak}=V_{pp}/2$) and also why equation 1.3.48 is correct for sinusoidal signals only.

Figure 7: Close-up of a t1_vibe_tra_130_1g] sequence.

To be compared with a sequence of square shaped pulses:

Figure 8: Close-up of an MPRAGE_3D sequence.
These signal may not look sinusoidal: but they are. The presented shapes are modulated sinusoidal signals which upon a much finer time scale can be resolved.
2 Materials and methods

2.1 The magnetic field probe

A simple magnetic field probe can be constructed by a loop of wire, or more conveniently a loop of coaxial cable. The latter of them also goes under the name: Faraday shielded coupling loop. The Wendelsten’s probe was made of an RG-223 coaxial cable with an impedance of 50Ω which had a break in the shield at half its circumference and the inner conductor shorted to the sheath at its neck, see Figure 9.

![Probe with inner conductor shorted to sheath at its neck and a break in the sheath at half its circumference.](image)

2.2 Calibration

A signal generator (Marconi TF2016, Marconi Instruments LTD, England) was used to generate an RF signal with appropriate frequencies. The signal was passed through an amplifier (ENI 3100L, Electronic Navigation Industries Inc, Rochester NY, USA) into a CF cell and terminated by a coaxial load resistor of 50Ω (BIRD 8135 Termaline, BIRD Electronic Corp., Cleveland Ohio, USA). A power meter (BRID 4431 Thruline, BIRD Electronic Corp., Cleveland Ohio, USA) was used to check if there were reflections returned through the CF cell after the terminating load. The calibration was done by placing the probe in the CF cell, aligning it so that the $H$-field polarization was along the central axis of the probe. Alignment of the probe described in Figure 9 was done ocularly, since a deviance of for example $5^\circ$ gives an error of less than 1%. An applied voltage, checked with an analogue voltage meter (HP 410C, Hewlett-Packard Company, Palo Alto California, USA), through the CF cell then generated a well known electric field obtained by simply taking the voltage divided by the distance between the top and septum plate. The relationship between the $E$- and $H$-field is described in Section 1.3.7. The
measured peak-to-peak voltage then had a simple relation to \( \mathbf{H} \) called the calibration factor, \( k_{\text{cal}} \), from equation 1.3.5. The calibration set-up is viewed in Figure 10.

Figure 10: A picture of the calibration set-up in the laboratory for association to Figure 1. Here the voltage meter and power meter are also shown.

The peak-to-peak voltage was measured for frequencies around 63.8 MHz which is the Larmor frequency for 1.5T. The CF cell dimensions enabled calibrations up to 90 MHz with good accuracy; however, above this threshold there would have been un-quantified uncertainties. Calibration for 3T (127.6 MHz) was not done for this reason. The calibration factor was acquired for a narrow interval of about ±1.5 MHz around the Larmor frequency, since the slice selection gradient, \( G_z \), in the MRI environment alters it slightly in addition to the field not being exactly 1.5T (or 3T). The highest possible voltage, namely 30 V, was used in the calibration to come as close as possible to MRI field strengths.

2.3 Experimental design inside the 1.5T MRI

The bore of the MRI had a diameter of 70 cm and was 125 cm long. \( X, y \) and \( z \) coordinates were defined in concordance with the MRI coordinate system with positive \( \hat{z} \) as the direction where patients are taken out from the MRI bore and \( \hat{y} \) upwards from the tray. Positive \( \hat{x} \) direction was then the cross product \( \hat{y} \times \hat{z} \). With the isocenter as a base, measurement points were selected with 80 mm spacing along the coordinate axes. These are marked in Figure 11 along with the coordinate system. Three tensor fields of rank zero, \( V_x(x) \), \( V_y(x) \) and \( V_z(x) \), available to hold information on peak-to-peak values from the Wendelsten’s probe obtained at location \( x \) could then be defined. The total tensor field sizes
were $7 \times 4 \times 13$, however, as a consequence of the bore having a circular cross-section along the z-axis, some elements in the tensor were simply set to *void* since their corresponding points were physically impossible to make measurements in.

![Coordinate System and Measurements Points](image)

Figure 11: Schematic view of the coordinate system and measurements points (black dots) inside the MRI bore seen from the z-axis and from the side.

A picture of the experimental set-up is viewed in Figure 12. The measurement base consisted of four sheets of 80mm thick styrofoam, hence each added sheet represented a new y-position. For faster data acquisition the probe positions were simply drawn as a grid on the uppermost sheet. The probe could then be positioned according to the grid and rotated to acquire three measurements for each position, one for every component ($V_{ptp,x}, V_{ptp,y}, V_{ptp,z}$). The probe was connected to a digital oscilloscope (Picoscope 5204) by an RG-223 double shielded coaxial cable passed through a high-pass filter with 10 MHz cut-off frequency to avoid influence from gradient fields. Data was passed into the Picoscope 6 software for visual read-out and direct transfer into a data-acquisition algorithm constructed with Matlab. Three measurements were done at each position for every point until the three tensors were completely filled. A total of around a thousand measurements including test measurements were made.
Data acquisition was made on a True Fast Imaging with steady state free precession (TRUFI)-sequence with one slice, and hence also only a single frequency to avoid frequency dependence of the probe as mentioned in Section 2.2. Some other sequences were tested, however, the most stable and reliable read-out was obtained with the TRUFI-sequence. The read-out was merely a voltage peak-to-peak value, corresponding to a particular magnetic field strength $H$ by the calibration factor in equation 1.3.5. Two sequences, the TRUFI and a Gradient Echo (GRE) sequence, were initially compared by measurements in a smaller scalar field. As expected, they displayed equal geometrical distribution after normalization ruling out sequence-dependent spatial variations. Some emphasis should be given to this topic, however, since it is not safe to assume that even though the MRI-scanner geometry is static: it is not necessarily the only factor invoking spatial variations of the RF-field when it comes to tissue interactions. MRI SMF ($B_0$) strength dictating the Larmor frequency and hence also resonant modes of metallic implants in patients or dimensions of receive coils are some examples of such additional factors (Kangarlu et al., 2005). For the in-air measurements made in this work there are presumably no other significant factors affecting spatial variations of the RF-field.

According to the manual the BC was no-tune, which means that it did not calibrate itself with patient or phantom dependent impedance matching. This was also observed in two ways: the characteristic
reflection artifacts in images acquired with a small phantom (approximately 1 liter or water) and measurement agreement when using different phantoms. This is of course important to know if results should be related to clinical practice.

2.4 Measurements on 3 T MRI

The idea of making these measurements was to get a hint on whether the spatial distributions of RF magnetic field amplitudes varied significantly between MRI scanners and SMF strength. No calibration was done for 3T, hence the obtained data was raw measurement values. Due to demands of the 3T MRI scanner (General Electric MR750, GE Healthcare, Little Chalfont, United Kingdom) the water phantom had to be a bit larger than the one used for 1.5T. Although a Steady-state Free Precession (SSFP) sequence with a little bit longer TR than the TRUF1-sequence was selected, equally reliable measurements were obtained. The measurements were in the absence of sufficient time at the 3T-lab carried out qualitatively, i.e. merely at 15 positions and along 3 dimensions with 150 mm spacing in the $z = 0$ plane.
3 Results

3.1 Calibration

The calibration for 1.5T was made in an interval between 62.5 and 65 MHz, to account for possible variations in Larmor frequency. Since the eigenfrequencies of the probe were unknown, this was important to rule out output spikes around the Larmor frequency. Some other frequencies (not visually presented here) were measured qualitatively and revealed a local frequency maximum at around 61 MHz.

![Peak-to-peak voltages for calibration around 63.8 MHz.](image)

Figure 13: Peak-to-peak voltages for calibration around 63.8 MHz.

A constancy test of the probe was also made in order to study input voltage dependence of $k_{cal}$. The expected result was a constant calibration factor according to equation 1.3.5 (or a linear increase in probe output). Figure 14 indicates a fairly stable calibration factor for input voltages above 15V.

![Graph of calibration factor vs. input voltage.](image)
Figure 14: Calibration factors for different input voltages.

Figure 15 displays the calibration factor, $k_{cal}$, as a function of frequency around 63.8 MHz for an input voltage of 30 V. The plotted interval is the same as in Figure 13 and the calibration factor reached a local minimum around 61 MHz to begin rising again at even lower frequencies, representing frequency sensitivity; however, in the narrow interval around 63.8 MHz the calibration factor was quite stable.

![Graph showing calibration factors around 63.8 MHz](image)

Figure 15: Calibration factors around 63.8 Mhz.

Initial measurements on the 1.5T MRI indicated TRUFI frequencies of around 63.6 MHz, corresponding to a calibration factor of

$$k_{cal} = 0.32 \text{ AV}^{-1} \text{ m}^{-1}$$
3.2 Measurements on 1.5 T MRI

A correspondence measurement of two sequences, TRUFI and GRE, was made to rule out any sequence dependent magnetic field strength variations. The measurements were done for the y-component at random positions with varying flip angle and can be seen in Figure 16.

![Figure 16: Correspondence of a TRUFI and GRE sequence at different locations](image)

Also an attempted measurement designed to correct for possible mismatch between the ideal geometrical midpoint (original measurement point (OMP)) and effective measurement point (EMP) of the probe was made. The measurement was done in x-direction for the z-component with assumed spatial symmetry of the field along the x-dimension for (y,z) = (0,24) cm. Measurements before and after correction are presented in Figure 17.

![Figure 17: EMP shift calculated from the symmetry assumption](image)

The EMP seemed to be shifted 0.8 centimeters towards the neck of the probe. In general, radiation intensity decreases with the distance, $r$, as $1/r^2$. Furthermore the BCC geometry was symmetric along...
the cartesian axes, hence the EMP could be calculated by fitting a $2^{nd}$ order polynomial to the OMP data and differentiating it, finding the minimum value and making the correction accordingly.

Results from exposure measurements are in this section presented in a variety of slices obtained by using the `interp3` and `slice` functions in MATLAB. For convenience a cylindrical coordinate system was used to obtain disk and tube slices. In current MRI EMF research cylinders are often the geometry at hand regarding specifically water phantoms. The algorithm used to pick the plane slices could actually have been used to draw any 3D-slice, preferably within the volume. Figures 18 to 24 show disk slices through different $z$-planes. Additional slices in the $z$-direction can be found in Appendix C. A very important note to remember when looking at each figure is that the fields are merely an at the surface representation and not in any way simulated or re-calculated depending on for example human geometry with varying conductive and dielectric properties.
Figure 18: 34 cm diameter disk slices at z=-40 cm from isocenter

The fields were not very strong at all for z=-40 cm. Upon increased proximity to the end rings (z=-24 cm) the fields dominated along the z-dimension.

Figure 19: 34 cm diameter disk slices at z=-24 cm from isocenter
In the vicinity of a common FOV (z=-8 cm) the fields were polarized mainly in the radial directions (x and y). In the z=0 plane there were almost no contribution in the z-dimension and the radial components (x and y) were fairly equal.

Figure 20: 34 cm diameter disk slices at z=-8 cm from isocenter

Figure 21: 34 cm diameter disk slices in the z=0 plane
Figure 22: 34 cm diameter disk slices at z=8 cm from isocenter

Contributions in the z=8 cm plane were similar to those in the z=-8 cm plane. The same holds for the z=24 cm plane as well and symmetry around the z=0 cm plane was obvious.

Figure 23: 34 cm diameter disk slices at z=24 cm from isocenter
Figure 24: 34 cm diameter disk slices at z=40 cm from isocenter.

Again, the fields 40 cm from the z=0 plane are small.
As can be seen in Figure 21 the x- and y-components of $\mathbf{H}_1$ were fairly equal in the $z=0$ cm plane. The dominating maximum component was rapidly switched from laying in the xy-plane at isocenter to pointing in the z-direction at -24 and 24 cm from the isocenter. The shown results are on 34 cm diameter disks, hence the difference was even more pronounced at x and y positions of for example ±24 cm. Axial symmetry could be sensed as well as symmetry around the $z = 0$-plane. It should be noted that all data were *maximum amplitude* components of $\mathbf{H}_1$, obtained without any reference time. The results are therefore merely presented as scalar fields rather than vectors.

A common geometry used in all kinds of dose calculation or measurement is the cylinder. It was therefore convenient to know also the fields at the surface of such a cylinder. This is presented in Figures 25 - 27.

![Figure 25: Cylindrical 34 cm diameter slice of the x-component.](image)

The cylindrical distribution of the x and y components were fairly equal.
Results

The z-component had its absolute strongest areas in the proximity of the end rings.

3.2.1 Protocol parameters

The measured spatial distribution of the TRUFI-sequence was applied to several other isocenter measured sequences which were part of clinical protocols. Some of the obtained parameters could then be recalculated for the entire protocol through exposure time weighting (time weighted averages). The most interesting parameters are presented in Table 2 both for the sequences and the protocols. A corresponding selection of sequence shapes are presented in Figures 41-42, whilst a complete set of sequence shapes are in Appendix B.
Table 2: Tested protocols and protocol parameters. *in y-direction at isocenter. **anywhere in the MRI bore in any direction.

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<th>TE / ms</th>
<th>Flip angle / °</th>
<th>Duration / mm:ss</th>
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<th>given B&lt;sub&gt;1,RMS&lt;/sub&gt; / %</th>
<th>measured B&lt;sub&gt;1,RMS&lt;/sub&gt; / µT</th>
<th>H&lt;sub&gt;1,peak&lt;/sub&gt; / Am−1</th>
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4 The values presented for each protocol (not sequences) are time-weighted averages of included sequences according to the Duration-column.
5 The given %-value indicates that the calculation of given B<sub>1,RMS</sub> might be a company secret; however, the corresponding 100%-value can be retrieved and will be shown momentarily.
6 Measured B<sub>1,RMS</sub> and H<sub>1,peak</sub> are related to H<sub>1,RMS</sub> and B<sub>1,peak</sub> according to equation 1.3.45 but presented in this more interpretable way. The H<sub>1,peak</sub> values can be multiplied with \( \mu = 4\pi \times 10^{-7} \approx 1.2610^{-6} \) to obtain B<sub>1,peak</sub> in µT.
Table 2 contains some valuable parameters which to some extent were taken directly from the MRI software. This included TR, TE, flip angle, duration, SAR and given $B_{1,RMS}$. SAR and given $B_{1,RMS}$ are control panel tools required by the IEC standards (International Electrotechnical Commission (IEC), 2010). Given $B_{1,RMS}$ serves as a complement to SAR which the operator should pay attention to when scanning patients with metallic implants where the body mass dependent SAR-value becomes insufficient. The remaining quantities are readouts and calculations done from the measurements. Sundström (2012) saw a possible linear relationship between the given SAR and measured $B_{1,RMS}$ squared (on a 3T GE MR750, GE healthcare, Erlangen, Germany). This theory was expected from equation 1.3.47 and gains further support here. Figure 28 presents this relationship for 1.5T together with a linear fit and corresponding equation.

![Figure 28: Plot of SAR versus (measured $B_{1,RMS}$)$^2$](image)

Solving the equation in Figure 28 for a SAR of 2 W/kg from Table 1 gives a $B_{1,RMS}$ of 0.94 μT. A similar attempt has been made done to correlate given $B_{1,RMS}$ with measured $B_{1,RMS}$. Given $B_{1,RMS}$ was presented as a percentage by the MRI software because it seemed the exact value is kept secret by the manufacturer. A linear relationship between the two is presented in Figure 29.
A correlation between given $B_{1,RMS}$ and measured $B_{1,RMS}$ was apparent. Given $B_{1,RMS}$ may be recalculated from the linear equation given in Figure 29 to given $B_{1,RMS}|_{100\%} = 1.40\mu T$ which is not equal to 0.94 obtained from the attempted SAR correlation. This indicates that given SAR and $B_{1,RMS}$ were separated alerts, the latter patient-independent as it should be.

The Picoscope had a function for determining the actual RMS-value of the signal. This was used to obtain measured $B_{1,RMS}$ by multiplying the read value with $\mu k_{cal}$. According to the IEC standards for medical equipment $B_{1,RMS}$ should be determined at isocenter. $H_{1,peak}$ was, however, not determined in the isocenter but obtained in a similar fashion by using the value of $V_{ptp}$, measured at the isocenter in y-direction, and multiplying it with 3.0 and $k_{cal}=0.32$. 3.0 was the ratio of the maximum of all spatial TRUFI measurements to an isocentric y-direction TRUFI measurement. This is an illustration of how the scalable tensors of spatial RF magnetic field strength measurements were used to extract information at other locations than the isocenter by only one isocentric measurement. Below is an extract of sequence shapes from Appendix B.
The big difference in sequence shapes were very clear between for example diffusion weighted EPIs and T1-weighted GREs. Notice also that the time and voltage axes are not identical in the two Figures.

3.2.2 Short notice on data acquisition

Regarding data acquisition three readouts were made. Two of them, PtP voltage and RMS voltage, were built-in functions of the Picoscope. For acquisition of the $D_{Cr}$ the mathematics channel-tool of the Picoscope was used. It lets the user enter any function calculative from any of the built-in functions or other user-constructed functions. Equation 1.3.48 was entered in this mathematics channel to acquire $D_{Cr}$. Relationships between $V$, $H$ and $B$ are described in sections 1.3.2 and 1.3.7.

3.2.3 Exposure at the surface of a human model

It is now maybe even more clear which of the $H_{1}$-components that dominated across different slices. Again the x- and y-components were fairly, but not entirely, equal depending on where the measurement point was. In a cylindrical coordinate system the radial component should be equal at equal radial distances for a given $z$ as described in Section 1.3. It was considered instructive to translate obtained knowledge of spatial variations of the RF magnetic field to a human model, which is not perfectly cylindrical. Surface human body meshes for MATLAB from NEVA Electromagnetics were used for this\textsuperscript{7}.

NB! A very important note is that these figures (32-34) are not in any way related to exposure within a human model but simply an in-air slice segment of a human surface.

Figure 32: x-component of the RF magnetic field at the surface of a human body model.

The x-component distributed quite evenly over the human surface. The y-component was not as strong, which was a consequence of the more elliptical geometry of a human.

Figure 33: y-component of the RF magnetic field at the surface of a human body model.

The upper arms and thighs could experience strong z-components since these regions may be in proximity of the end rings while performing certain scans.
Figure 34: z-component of the RF magnetic field at the surface of a human body model. Note particularly the more intense spots around the arms and thighs.

The probe was made of coaxial cable and hence hard to bend, which was why values outside ±24 cm could not be obtained. The highest field strengths on the human body surface were measured at the upper arms and thighs. Of course this depends on where a patient is located, but could for example be the case in liver, lung, spinal cord or cardiac scans etc. Differences between x and y components are explained by the “elliptical” cross-section of the body and that contributions from the end rings are more pronounced in the x than y-direction as we move along it and vice-versa. Equations 1.3.20 to 1.3.23 may serve as good help to understand why this is. The following three Figures (35-37) are slices through the y=0 plane with x, y and z components measured respectively.

Figure 35: x-component of a y=0 slice in the 1.5T MRI. Presented data are pure $V_{p/p}$ values interpolated on a mesh with 5x5 mm element size.

A complete slice of the x-component reveals that contributions from the end rings showed in the x-component for the y=0 plane and not in the y-component. The reversed scenario would have been seen for a slice in the x=0 plane.
Results

Figure 36: y-component of a y=0 slice in the 1.5T MRI. Presented data are pure $V_{PP}$ values interpolated on a mesh with 5x5 mm element size.

A complete slice of the z-component revealed a big dominance around the end rings and very low contributions in a common FOV.

Figure 37: z-component of a y=0 slice in the 1.5T MRI. Presented data are pure $V_{PP}$ values interpolated on a mesh with 5x5 mm element size.

These Figures should be used as comparison to the 3T measurements in the upcoming Section 3.3.

3.3 Measurements on 3 T MRI

The 3T measurements (Figures 38-40) were not at all as detailed as those for 1.5T due to accessibility issues. The purpose of these measurements were to give hints on whether the obtained spatial distribution for 1.5T seemed reasonable, or if there were any significant differences. Overall the $V_{PP}$ output to the oscilloscope was lower than for 1.5T. It was; however, not certain that this also meant a smaller $B_1$-field since no calibration for 127.6 MHz could be made.
Figure 38: x-component of a y=0 slice in the 3T MRI. Presented data are pure $V_{PtP}$ values interpolated on a mesh with 2x2 mm element size.

Similar distributions as for 1.5T could be sensed for the x- and y-component. Lack of density in the measurements provide little support and skepticism to the interpolation should be held.

Figure 39: y-component of a y=0 slice in the 3T MRI. Presented data are pure $V_{PtP}$ values interpolated on a mesh with 2x2 mm element size.

As for 1.5T the strongest z-components seemed to be located in proximity of end rings; however, exact location of the end rings were unknown.
The GRE sequence used for 3T was as reliable as the TRUFI sequence for 1.5T. Generally, regarding sequence shapes, the pulses seemed to have longer duration compared to 1.5T. No $B_{1,RMS}$, $H_{1,peak}$ or DC values were measured for 3T.

### 3.4 Association to numerical simulation

A consequence of the results in the two previous sections is that RF field strength seems to be attributable to a specific machine in a specific set-up; that is: coil design parameters like size, number of rungs, resonance frequency, capacitances etc. and what is placed inside the coil. To date a literary review indicated that numerical simulations on a Siemens Magnetom Espree 1.5T with the exact same specifications as the one employed here was non-existent. There are, however, studies made on other MRI-like models (Gurler and Ider, 2012; Ibrahim et al., 2000, 2005) that can be quantitatively associated with the results obtained here.
4 Discussion

4.1 Regarding the method used

An important component in the initial planning was to determine a sufficient number of measurement points. Several factors had to be accounted for when choosing a suitable measurement accuracy, the largest of them being time and reliability of the result. As an intermediate step a test measurement of 21 measurements was timed for further planning. The acquisition time of the result was highly dependent on spacing between measurement points. An initial wish to obtain tensors with 50 mm point spacing had to be redesigned to tensors with 80 mm point spacing which effectively lowered the total acquisition time by approximately 70% without causing too much of an impact on “hot-spot” coverage of the RF magnetic field strengths. Initial measurements gave indications of very fast switching x/y/z component dominance of total magnetic field strength, hence calling for minimized point spacing.

An attempt to outline not only spatial variations of RF magnetic field amplitude but variations of RF phase as well was made. It appeared that calculation of a resultant vector field was impossible to carry out without knowledge of the RF phase. The idea of using a reference probe was tested and actually worked in the sense of giving a representation of the RF phase. The phase at different positions in space did however depend on two parameters: time shift from a reference time and location in space, since the fields propagate at the speed of light. Also, the attempted solution gave no information on whether the phase was for example $+\phi$ or $\phi - 2\pi$ at an arbitrary point. Knowledge of this was obtained after a majority of measurements were already completed, whereas it would have required a hefty amount of extra time to assess. Furthermore, the instruments at hand were not built to work as a vector network analyzer (VNA) but could potentially do so with some external software implementation. There are, however, plenty of such commercially available instruments on the market. At one hand it is tempting to propose that a cylindrical coordinate system would have been a better approach during measurements; that is: that radial, axis of rotation and z-components were measured instead of x-, y- and z-components. On the other hand this would have implied that knowledge of end ring contributions had been harder to see. Either way measurements such as these are very time consuming wherefore a recommendation to anyone tempted to perform similar measurements is to use a VNA.

4.2 Regarding the results

Overall, the findings in this Master’s thesis were quite expected as far as spatial distribution is considered. COMSOL simulations made by Gurler and Ider (2012) show good agreement with the mea-
measurements distribution-wise and supposedly also field strength-wise. It should be noted that these simulations were for a BCC with a diameter of 10 cm compared to 70 cm for the Espree Slim 70 BC; however, they expected a maximum field strength of around 22-23 A/m which is not that far from the maximum of \(\sim 33\ \text{A/m}\) measured here.

Correspondence between SAR and measured \(B_{1,RMS}^2\) proposed by Sundström (2012) was reproduced in this work (Figure 28). A stronger relationship between given \(B_{1,RMS}\) and measured \(B_{1,RMS}\) was found (Figure 29). Reliability for the exposure alert in MRI is thus established and the recalculated value of given \(B_{1,RMS}\) is determined to 1.4\(\mu T\). Hence given SAR and \(B_{1,RMS}\) are simple tools accessible for the operator in clinical practice and when performing exposure studies aimed at temperature rise during MRI scans. An indirect finding following these observations is that neither can \(H_{1,peak}\) nor DC be directly associated to any of the given parameters from the MRI control panel. As mentioned in Section 1 studies on genotoxic effects have been contradictory (Lee et al. (2011) versus Szerencsi et al. (2013)) even when all exposure parameters have seemed to be in agreement. An implication of the correctness in SAR and given \(B_{1,RMS}\) is that none of them are sufficient to explain contradictory research on genotoxicity; however, they are sufficient to describe temperature rise. The results in this Master’s thesis imply that \(H_{1,peak}\) is a hidden parameter in to date research on genotoxic effects in MRI and suggests that particularly \(H_{1,peak}\) and maybe also DC may be needed in a future dosimetric model of EMF exposure in MRI. It is also suggested that \(B_{1,RMS}\) and SAR may be weakly linked to genotoxic effects and that there is a stronger link to another parameter. It was desired for this Master’s thesis to lay some ground work in the process of parameterizing MRI sequences for a more clear classification in cell-biological exposure studies. If the reader takes a look in Appendix B it is clear that multiple sequence shapes differ considerably. The sequence shapes could possibly be a target for classification, but on the other hand there might be too much variation going on across different systems which render this approach impractical in the long run, that is: if a cohort study should be made. Nonetheless it is very simple to make an in-air exposure measurement with a magnetic field probe to obtain a sequence shape.

The measurements were in-air measurements. If the measurements were done in for example water the relation between wavelength/probe diameter changes drastically as described by Hoult and Phil (2000) due to changes in speed of propagation. This does not affect the magnetic field strength directly but may introduce sources of resonance in human subjects which indirectly induces hot-spots. Such hot spots have been observed in numerous simulations (Murbach et al., 2011, 2013; Nadobny et al., 2007; Neufeld et al., 2011). Some authors, like Watanabe (2012), have highlighted the position dependency of magnetic field strength due to the phase change along the laboratory frame in the x and y directions. Actually this is true for the z-direction as well since we have contributions from the end rings sometimes
not taken into account in the literature. It is interesting to see the correspondence of hot-spots in arms found by Murbach et al. (2011) and Neufeld et al. (2011) to spatial RF magnetic field distribution measured in this thesis. According to Faraday’s law of induction the surface of the arm also forms a suitable geometry for resonance. The fields at these positions do clearly have a great z-component which might yield unfavorable situations when conducting objects form large cross-sectional areas in the vicinity of the end rings. This is interesting from a cell-biological exposure point of view. Luckily limbs are in general relatively insensitive to radiation damage; however, a classification of MRI protocols and sequences are important, but what is at least as important is consistent and correct experimental set-ups. This addresses specifically varying conditions when exposed samples are taken in-vitro, ex-vivo or in-vivo since geometries of test tubes, petri dishes and humans differ. Eddy currents, hot-spots and proper SAR depend on geometry and effects of resonance in certain geometries may affect the result considerably. Of course the same thing holds for positioning the sample since the RF magnetic fields in MRI propagate in three dimensions thus affecting the cross sectional area of any object as seen from the direction of polarization in that particular point. Large variations of actual exposure is proposed between isocentric exposures and exposures in proximity of the end rings.

Regarding future research it is concluded that numerical simulations are verified if they are well designed. Time consuming measurements like these are not recommended on a larger scale but in some cases it can definitely be a valuable tool to verify that developed FDTD or FEM models work as they should and that nothing has been missed. In most cases two measurements might be enough to validate a simulation model to sufficient accuracy. This applies not only to exposure research but to image acquisition and quality research as well.
5 Conclusions

It was concluded that large spatial variations in the RF magnetic field exist within the MRI bore. Cell-biological studies should be made in the largest field strengths which are in the periphery of the coil, in proximity of the end rings. Upon approaching the end rings the dominating RF magnetic field component switches from transverse direction to longitudinal direction.

Large differences in $H_{1,\text{peak}}$ through a set of sequences were seen. These differences did not depend on variations in RMS or SAR. Two sequences with similar RMS or SAR values may show up to a 400% difference in peak values.

Exposure research on genotoxicity, which to date supposedly contains hidden parameters, may be deceived by the coupling of $B_{1,\text{RMS}}$ to SAR and tissue heating. It might be simple to interpret the SAR value as a dosimetric quantity attributed to genotoxicity even though that might be an incomplete description. The complete interaction mechanisms between RF magnetic fields and tissue remain inconclusive; however, the RF magnetic field peak value is an interesting hidden parameter in to date research. Development of consistent exposure conditions with the starting point in thorough parameterization of sequences is suggested.
References


Appendices

A Derivations

Derivation of equation 1.3.20: Consider the vector potential for a straight segment of wire. Let \(z_1\) and \(z_2\) be the distances to the two endpoints of the wire segment and \(s\) the distance between the source to the field point. Let \(s\) be the right angle distance between the wire and the point and \(z\) the distance from the right angle point to the field point. Then \(r = s^2 + z^2\). Also \(A(r) \equiv A\), hence the \(r\)-dependence is understood.

\[
A = \frac{\mu_0 I}{4\pi} \int_{z_1}^{z_2} \hat{z} \frac{\hat{r}}{\sqrt{\mathbf{s}^2 + \mathbf{z}^2}} dz = \frac{\mu_0 I}{4\pi} \hat{z} \left[ \ln \left( z + \sqrt{s^2 + z^2} \right) \right]_{z_1}^{z_2} = \frac{\mu_0 I}{4\pi} \hat{z} \left[ \ln \left( z_2 + \sqrt{s^2 + z_2^2} \right) \right]_{z_1} \quad \text{By remembering that } B = \nabla \times A.
\]

One finds

\[
B = \frac{\mu_0 I}{4\pi} \left[ \ln \left( z_2 + \sqrt{s^2 + z_2^2} \right) \right] \frac{\hat{x}}{\ln (z_1 + \sqrt{s^2 + z_1^2})} = \frac{\mu_0 I}{4\pi} \left( \frac{\partial}{\partial y} \hat{x} - \frac{\partial}{\partial x} \hat{y} \right) \left[ \ln \left( z_2 + \sqrt{s^2 + z_2^2} \right) \right] \frac{\hat{x}}{\ln (z_1 + \sqrt{s^2 + z_1^2})} = \frac{\mu_0 I}{4\pi} \left( \frac{\partial}{\partial y} \hat{x} - \frac{\partial}{\partial x} \hat{y} \right) = \frac{\mu_0 I}{4\pi} \left( \frac{\partial}{\partial y} \hat{x} - \frac{\partial}{\partial x} \hat{y} \right) = \frac{\mu_0 I}{4\pi} \left( \frac{\partial}{\partial y} \hat{x} - \frac{\partial}{\partial x} \hat{y} \right) = \frac{\mu_0 I}{4\pi} \left( \frac{\partial}{\partial y} \hat{x} - \frac{\partial}{\partial x} \hat{y} \right) = \frac{\mu_0 I}{4\pi} \left( \frac{\partial}{\partial y} \hat{x} - \frac{\partial}{\partial x} \hat{y} \right) = \frac{\mu_0 I}{4\pi} \left( \frac{\partial}{\partial y} \hat{x} - \frac{\partial}{\partial x} \hat{y} \right) (A.0.1)
\]

The inner derivatives of \(s(x,y)\) are easily found by realizing that

\[
s = \|s\| = \|(x - R \cos (\Delta \phi))\hat{x} + (y - R \sin (\Delta \phi))\hat{y}\| = \sqrt{x^2 + y^2 - 2xR \cos (\Delta \phi) - 2yR \sin (\Delta \phi)}
\]
then
\[
\frac{\partial s}{\partial y} = \frac{2y - 2R \sin (\Delta \phi)}{2 \sqrt{x^2 + y^2 + R^2 - 2xR \cos (\Delta \phi) - 2yR \sin (\Delta \phi)}} = \frac{y - R \sin (\Delta \phi)}{s}
\]
\[
\frac{\partial s}{\partial x} = \frac{2x - 2R \cos (\Delta \phi)}{2 \sqrt{x^2 + y^2 + R^2 - 2xR \cos (\Delta \phi) - 2yR \sin (\Delta \phi)}} = \frac{x - R \sin (\Delta \phi)}{s}
\]

Its convenient to note that if the length of a wire is \( L \) the distances \( z_1 \) and \( z_2 \) to each end of the wire at position \( z \) will be either \( z \) or \( L - z \). One may also freely choose a coordinate system along the \( z \)-axis (although it needs to be kept throughout the calculations of course). Finally, substituting the derivatives, the expression for \( s, z_1 = z \) and \( z_2 = L - z \) into the final expression of equation A.0.1 one obtains equation 1.3.20.

**Derivation of equations 1.3.21-1.3.23.** Start with the Biot-Savart law (equation 1.3.15). Here the separation vector is \( \mathbf{s} = (x - R \cos \phi)\mathbf{x} + (y - R \sin \phi)\mathbf{y} + z\mathbf{z} \) and the coordinates to a source segment of the circular loop \((x', y', z') = (R \cos \phi, R \sin \phi, 0)\). \( z' = 0 \) comes from the ability to freely choose a coordinate system. For convenience it is placed in the same plane as the loop (or end ring), and the compensation for this is that \( z \), the axial distance to the field point, takes another value. From this one finds \( d\mathbf{l}' = (dx', dy', dz') = (-R \sin \phi, R \cos \phi, 0)d\phi \). Evaluating the cross product:

\[
d\mathbf{l}' \times \mathbf{s} = \begin{vmatrix} \mathbf{x} & \mathbf{y} & \mathbf{z} \\ -R \sin \phi d\phi & R \cos \phi d\phi & 0 \\ (x - R \cos \phi) & (y - R \sin \phi) & z \end{vmatrix} = Rz \cos \phi d\phi \mathbf{x} + Rz \sin \phi d\phi \mathbf{y} + (R^2 - Rx \cos \phi - Ry \sin \phi)d\phi \mathbf{z}
\]

Continuing one finds \( s^3 = (x^2 + y^2 + z^2 + R^2 - 2Rx \cos \phi - 2Ry \sin \phi)^{3/2} \). Plugging this information into the Biot-Savart law one correctly obtains equations 1.3.21-1.3.23 expressed component wise.
B  Sequence shapes

Figure 41: ep2d_diff_tra_b0_b800_p2_160

Figure 42: GD_MPRAGE

Figure 43: gre_angio
SPATIAL VARIATION OF RADIO FREQUENCY MAGNETIC FIELD EXPOSURE FROM CLINICAL PULSE SEQUENCES IN 1.5T MRI

Figure 44: pd_tse_cor_fs_384

Figure 45: MPRAGE_3D

Figure 46: t1_tse_fs_tra_p2
Figure 50: t1_vibe_fs_tra
C Disk slices

Figure 51: 34 cm diameter disk slices at z=-48 cm from isocenter

Figure 52: 34 cm diameter disk slices at z=-32 cm from isocenter
Figure 53: 34 cm diameter disk slices at z=-16 cm from isocenter

Figure 54: 34 cm diameter disk slices at z=16 cm from isocenter
SPATIAL VARIATION OF RADIO FREQUENCY MAGNETIC FIELD EXPOSURE FROM CLINICAL PULSE SEQUENCES IN 1.5T MRI

Figure 55: 34 cm diameter disk slices at z=32 cm from isocenter

Figure 56: 34 cm diameter disk slices at z=48 cm from isocenter