Bridging Human Responses and Imaging Performance in 7 Tesla MRI - Balancing Safety, Comfort, and Image Quality

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Abstract:

Background: Ultra-high field (UHF) 7 Tesla magnetic resonance imaging (MRI) offers significant improvements in signal-to-noise ratio and spatial resolution. However, it also presents challenges such as increased acoustic noise, peripheral nerve stimulation (PNS), and various potential physiological and perceptual effects. Therefore, from a human response perspective, it is crucial to consider both safety and participant comfort alongside image quality in UHF MRI. This balance is essential for the safe and effective implementation of UHF imaging.

Aim: The overall aim of this thesis was to investigate the physiological and perceptual effects associated with the 7 T MRI environment, including cognitive performance, auditory function, acoustic noise, and PNS, as well as to evaluate technical strategies for reducing these effects while preserving image quality.

Method: Across four studies, cognitive performance, auditory function, and perceptual responses were investigated in healthy participants. Furthermore, the effects of acoustic noise reduction strategies, including the gradient modulation technique SofTone, were evaluated in terms of both safety-related outcomes and quantitative and qualitative measures of imaging performance.

Results: The results indicate that 7 T MRI may be associated with transient sensory and perceptual effects, while cognitive performance remains stable. Effective dual hearing protection appears to preserve outer hair cell function despite repeated exposure to high acoustic noise levels. Acoustic noise reduction techniques significantly reduce sound pressure levels and PNS but may introduce trade-offs in image quality depending on the type of sequence used.

Conclusion: Taken together, the findings highlight the importance of balancing technical optimisation with physiological and perceptual safety in UHF MRI. The thesis underscores the need for a comprehensive approach that integrates human experience with image quality metrics to support both participant comfort and diagnostic performance in UHF MRI.

Keywords: Magnetic Resonance Imaging, Ultrahigh Magnetic Field, Hearing, Ear Protective Devices, Noise, Cognitive Function, Image Quality, Healthy Volunteers

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Original papers

This thesis is based on the following studies, which have been reprinted with permission from the publishers. Throughout the thesis, the four studies are referred to with Roman numerals.

I. Wennberg, L., Mårtensson, J., Langensee, L., Sundgren, P.C., Markenroth Bloch, K., Hansson, B. *Effects of ultra-high field MRI environment on cognitive performance in healthy participants.* Radiography (Lond). 2023;30(1):95-9.

II. Wennberg, L., Waechter, S., Brännström, K.J., Sundgren, P.C., Markenroth Bloch, K., Hansson, B., Mårtensson, J. *Dual hearing protection effectively preserves outer hair cell function during 7 T MRI.* Radiography (Lond). 2026;32(4):103398.

III. Glans, A.[*], Wennberg, L.[*], Wilén, J., Lindgren, L., Sundgren, P.C, Mårtensson, J., Markenroth Bloch, K., Hansson, B. *Evaluation of Software-Optimized Protocols for Acoustic Noise Reduction During Brain MRI at 7 Tesla.* J Magn Reson Imaging. 2025;62(2):577-87.

IV. Wennberg, L., Mårtensson, J., Sundgren, P.C., Markenroth Bloch, K., Rumetshofer, T., Hansson, B., Glans, A. *Acoustic Noise Reduction at 7 T MRI: Effects on Peripheral Nerve Stimulation and Quantitative Image Quality* (manuscript form).

[*]*Shared first authorship*

The author´s contribution to the papers

Papers I and II

The doctoral student contributed to study planning and was responsible for participant recruitment and inclusion, as well as scheduling MRI examinations following safety screening. The doctoral student performed most of the MRI examinations, cognitive testing, and hearing assessments. The doctoral student also participated in the setup and execution of sound level measurements of MRI sequences in collaboration with an audiologist. Furthermore, the doctoral student was responsible for compiling cognitive and hearing data, conducting statistical analyses, interpreting the results, and preparing figures and tables. The doctoral student contributed to drafting and revising the manuscripts and acted as the corresponding author during submission.

Papers III and IV

The doctoral student contributed to the study design and was responsible for participant recruitment, inclusion, and scheduling of MRI examinations. The doctoral student participated in and performed all MRI examinations, collecting data on peripheral nerve stimulation (PNS), acoustic noise measurements, and participants' subjective experiences related to PNS and noise.

In Paper III, the doctoral student coordinated qualitative image assessment and participated in all assessment sessions with the neuroradiologists to ensure randomisation and blinding of MRI sequences. The doctoral student also contributed to manuscript preparation and revision.

In Paper IV, the doctoral student was responsible for compiling data, conducting the quantitative image assessment and statistical analyses, interpreting results, preparing tables and figures, drafting the manuscript, and will be the corresponding author during submission.

Language editing and AI assistance

This thesis was written entirely by the author, who is responsible for all content and interpretations. ChatGPT was used solely for language editing to improve clarity and readability. It was also used to generate two schematic figures: a conceptual illustration of the radiographer's central role and an illustration of the data collection process for Paper I. Both figures were produced based on the author's instructions, ideas, and conceptual design. No AI tools were used in ways that could influence the scientific content, data analysis, interpretation of results, or conclusions.

Abbreviations

3D	three-dimensional
ANR	acoustic noise reduction
B_0	main static magnetic field
CNR	contrast-to-noise ratio
CSF	cerebrospinal fluid
dB / dBA	decibel / A-weighted decibel
dB/dt	rate of change of the gradient magnetic field
DSST	Digit Symbol Substitution Test
DPOAE	Distortion Product Otoacoustic Emissions
FDA	U.S. Food and Drug Administration
fMRI	functional magnetic resonance imaging
FOV	field of view
FSE	fast spin echo
ICNIRP	International Commission on Non-Ionizing Radiation Protection
IEC	International Electrotechnical Commission
MRI	magnetic resonance imaging
LAeq	A-weighted equivalent continuous sound level
LAFmax	maximum A-weighted sound level with Fast time weighting
LCpeak	C-weighted peak sound pressure level
OAE	otoacoustic emissions
PNS	peripheral nerve stimulation
ROI	region of interest
RF	radiofrequency
Rx	receive coil
SAR	specific absorption rate
SNR	signal-to-noise ratio
SPLs	sound pressure levels
T	Tesla
T1W	T1-weighted
T2W	T2-weighted
TE	echo time
TR	repetition time
TSE	turbo spin echo
Tx	transmit coil
UHF	ultra-high field

Preface

The world of magnetic resonance imaging (MRI) has fascinated me since the beginning of my career in radiology. At that time, ultra-high field scanners were not yet available, and 1.5 Tesla systems represented the clinical standard. Even then, I was drawn not only to the technical sophistication of imaging, but also to the human aspect of the experience—how patients perceive, respond to, and are affected by these examinations. These experiences sparked a curiosity that has guided my professional journey ever since.

Years later, when returning to the MRI unit as a radiographer at the 7 Tesla facility at Lund University Hospital, I entered a transformed environment. Technology had advanced in ways I had previously only imagined. At the same time, it raised new questions regarding how individuals experience the examination, how acoustic noise, the confined space, and the environment may influence cognitive and physiological responses, and whether high magnetic fields could have subtle effects. Answering these questions required not only a deeper understanding of the individual's experience and responses to the MRI environment, but also the development and evaluation of measures that could make MRI examinations safer and more tolerable. Sound-reduction methods were therefore evaluated as strategies to reduce discomfort and support individuals' well-being. Investigating these issues required a multidisciplinary approach, integrating expertise from physics, psychology, audiology, radiography, and medicine to examine both the technical and experiential dimensions.

A central motivation for this work has been to emphasise the radiographer's role in identifying clinical challenges and opportunities for improvement—by enhancing safety, refining the individual's experience, and supporting person-centred care. This perspective also reflects the broader evolution of radiography as a profession and its ongoing efforts to strengthen its scientific foundation.

This thesis represents both the culmination of a long-standing curiosity and a step forward in research, bridging clinical practice, individual experiences, and scientific inquiry. I hope these findings contribute to a deeper understanding of how individuals respond to the MRI environment and acoustic noise, inform safer and more person-centred imaging practices, and inspire continued multidisciplinary collaboration.

I am deeply grateful to my supervisors, Pia Maly Sundgren, Boel Hansson, Johan Mårtensson, and Karin Markenroth Bloch, for their invaluable guidance, and to my colleagues at Lund University Hospital and the Swedish 7 Tesla national facility for their support in data collection and stimulating discussions.

Introduction

Magnetic resonance imaging (MRI) provides high-resolution, non-invasive imaging without ionising radiation, but requires strict safety procedures governed by international standards and regulatory guidelines for both participants and staff.[1] Over time, MRI technology has progressed toward higher magnetic field strengths, with ultra-high field (UHF) systems such as 7 Tesla (T) offering exceptional image detail while introducing additional technical and physiological challenges. Understanding the MRI environment, the advantages and complexities of UHF imaging, and associated safety considerations is therefore essential. Radiographers play a key role in this process, integrating expertise, collaborating across disciplines, and applying strategies to maintain image quality while minimising potential adverse effects.

The MRI environment

A thorough understanding of the MRI environment is essential for ensuring participant and staff safety, as well as for managing potential physiological and perceptual effects.[1,2] MRI systems generate three principal electromagnetic fields: the static magnetic field, the switched gradient magnetic fields, and the radiofrequency (RF) field. These fields are involved in creating images,[3] but can also affect the human body in different ways.[4] Awareness of these effects is crucial for implementing appropriate safety procedures, optimising participant comfort, and minimising adverse responses during scanning.

The static magnetic field

The static magnetic field (B_0), generated by the main magnet of the MRI system, is the most distinctive feature of MRI and a primary source of risk. Due to its strength, which far exceeds magnetic fields encountered in everyday life, it exerts substantial translational and rotational forces on ferromagnetic objects, potentially resulting in serious safety hazards.[3] In addition to effects on ferromagnetic materials, exposure to the static field and the associated electromagnetic environment may also pose risks related to implanted or external devices, including device malfunction.[4,5] Furthermore, clinical superconducting magnets are typically kept continuously energised, meaning that the static magnetic field is always present. For this reason,

the scanner room is a strictly controlled environment, accessible only to authorised personnel with specialised MRI safety training.[1] These precautions are essential for protecting both staff and participants and for ensuring safe system operation.

Clinical MRI scanners typically operate at field strengths up to 3 T,[6,7] whereas UHF 7 T systems are primarily used in research settings.[8-11] Clinical implementation of 7 T MRI is still in its early stages, with an increasing number of scanners becoming available for patient care.[12-15]

MRI has been used extensively since the early 1980s, with no evidence of harm from static magnetic fields.[16] Nevertheless, physical movement within a static magnetic field gradient can induce sensations such as vertigo, nausea, phosphenes, and metallic taste at field strengths above 2-4 T.[4,17] Vertigo is associated with nystagmus, i.e., involuntary eye movements, as head rotation is linked to ocular gaze direction.[4] Moving patients slowly into the magnet bore can help to reduce movement-induced sensory effects.[18] An additional sensation related to vertigo is the perception of illusory motion during movement through the magnet bore. This may include a mismatch between perceived and actual trajectory, as well as a sense of rotation or curved motion while travelling through the scanner.[19-23]

Sensitivity to these effects varies considerably between individuals.[24] Static magnetic fields above approximately 0.1 T can also induce measurable flow potentials in major blood vessels, particularly around the heart. These findings are consistent with a World Health Organization (WHO) expert consensus report, which reflects international expert opinion but does not necessarily represent official WHO or International Electrotechnical Commission (ICNIRP) policy.[17,25] In contrast, reported injuries involving ferromagnetic objects or cardiac pacemakers highlight the importance of strict safety precautions.[16] Although clinical MRI systems typically operate at 1.5-3 T, providing a substantial safety margin, continued efforts are required to prevent accidents related to undetected ferromagnetic materials.[16,26]

The time-varying gradient field

The gradient system is a fundamental component of the MRI scanner and plays a crucial role in image formation by enabling spatial selection and signal encoding. MRI systems are equipped with gradient coils consisting of resistive windings that generate position-dependent variations in the magnetic field strength. During image acquisition, these gradient fields are rapidly switched on and off, typically in conjunction with radiofrequency excitation pulses, to encode spatial information in the emitted radiofrequency signal.[27]

The performance of the gradient system is commonly characterised by two key parameters: the maximum gradient amplitude, measured in mT/m, and the slew rate, which describes how quickly the gradient can reach a specified amplitude within a

given time (T/m/s). Together, these parameters determine how efficiently and rapidly spatial encoding can be achieved.[28]

The rapid switching of gradient fields produces time-varying magnetic fields and requires large electrical currents in the gradient coils. This process is also the primary source of the high levels of acoustic noise observed during MRI examinations. Acoustic noise during MRI can cause annoyance, interfere with verbal communication, and increase anxiety.[29,30] Furthermore, it has been demonstrated that there is a direct correlation between the acoustic noise level and claustrophobia.[31] Additionally, certain patient groups are particularly vulnerable to acoustic noise, including individuals with psychiatric disorders, the elderly, and children who may become confused or anxious. Sedated patients can experience discomfort, while some medications may increase hearing sensitivity.[32-34] Neonates are also considered especially vulnerable, as their auditory system is still under development.[35] Participants' perceptions of acoustic noise during MRI acquisitions can also vary considerably, ranging from mild annoyance to feelings of fear.[36] Furthermore, previous studies have shown that scanner acoustic noise during functional MRI (fMRI) can influence both auditory and non-auditory brain networks, modulating activation patterns and potentially reducing fMRI data quality.[37-39]

Additionally, the rapid changes in the resulting magnetic field at specific locations within the gradient can induce peripheral nerve or muscle stimulation (PNS).[40] When the induced electric field is sufficiently strong, it can depolarise neuronal membranes by altering the transmembrane potential, triggering action potentials in peripheral nerves.[4,6,41] This can lead to involuntary muscle contractions and sensory perceptions ranging from mild tingling to discomfort or even pain.[3,42-44] Only the most linear central portions of the magnetic gradient fields are used for imaging within the field of view (FOV). As a result, the largest gradient field strengths may occur outside the FOV, where linearity is not required for image formation.[45]

The PNS threshold is commonly expressed in terms of dB/dt, defined as the rate of change of the magnetic field over time, which directly influences the induced electric fields that can stimulate peripheral nerves.[46] Studies have shown that the intensity of dB/dt required to elicit an uncomfortable sensation is approximately 50% higher than the initial perception threshold.[47] This indicates a distinct margin between the onset of PNS and the level at which it becomes uncomfortable.

Cardiac stimulation is extremely unlikely in modern MRI due to safety limits, and PNS is generally not a concern. Nevertheless, care should be taken to minimise gradient-related discomfort in patients.[41]

The radio frequency field

In MRI, a coil generates the radiofrequency electromagnetic field (the "RF field"), which excites nuclear spins to produce signals. The magnetic component of the RF field (B_1) is much weaker than the main B_0 field but oscillates rapidly at the Larmor frequency. Despite its low strength, the high frequency of B_1 raises safety concerns.[3] The RF field used in MRI can induce currents in tissue, which may lead to resistive heating depending on tissue conductivity, geometry, and position relative to the excitation coil. The absorption of this RF energy by the body is quantified using the dosimetry term specific absorption rate (SAR), which provides a measure of energy deposited per unit mass of tissue.[48,49] SAR is the mass-normalised rate at which RF energy is deposited in biological tissue, typically expressed in watts per kilogram. Individual exposure during an MR procedure is usually described in terms of whole-body averaged SAR and local peak SAR (averaged over 1g of tissue).[50]

Although RF components are not the main source of MRI acoustic noise, vibrations within the system contribute to the sound experienced by the patient. Eddy currents induced in the RF coils and RF shield by rapidly switching gradient fields cause vibrations that, despite being relatively small, generate significant noise due to the coils' proximity to the patients' ears and bone conductivity.[51,52]

MRI at ultra-high field

The rapid development of UHF MRI, particularly 7 T, has opened new possibilities for neuroimaging. Higher magnetic field strengths lead to an increase in signal-to-noise ratio (SNR), which benefits a wide range of MRI applications. This additional SNR can be used to improve spatial resolution through smaller voxel sizes, enabling more detailed anatomical depiction. It can also be used to enhance temporal resolution in dynamic imaging, allowing faster physiological processes to be captured with greater fidelity. Furthermore, scan times can be reduced through undersampling strategies, enabling faster acquisition.[7,53-56]

A key driver in the development of higher MRI field strengths has been fMRI. The blood oxygen level-dependent (BOLD) signal arises from local changes in blood flow and volume associated with neuronal activation, mediated through multiple MRI contrast mechanisms. These effects exhibit a field-strength dependence, resulting in a superlinear increase in BOLD sensitivity with B_0 and an increased weighting toward microvascular contributions near sites of neuronal activity.[56-58]

These advantages support both clinical diagnostics and research applications. However, high operational complexity, installation costs, and the need for protocol refinement are likely to limit widespread use, restricting it to a small number of specialised centres where 7 T studies complement 1.5 T and 3 T examinations.[7,59]

Clinicians must also adapt to unprecedented contrasts, higher anatomical resolution, and develop new skills for accurate interpretation of these advanced images.[9]

UHF 7 T MRI systems are typically characterised by a longer and more enclosed bore compared with lower field strength systems, requiring the participant to be positioned deeper within the magnet and exposed to a more extensive magnetic environment throughout the examination. This spatial configuration may influence the overall patient experience, particularly in terms of confinement and perceived claustrophobia.[60] Previous research has demonstrated that, for patients scanned on both systems, the rate of claustrophobic reactions was significantly lower when using a short-bore scanner.[31]

Safety considerations in 7 T MRI systems

Current international safety guidelines indicate that no evidence of serious acute health effects has been observed for static magnetic field exposure up to 8 T, as stated by ICNIRP.[25] In line with this, the International Electrotechnical Commission (IEC) increased the static magnetic field limit for the first level controlled operating mode from 4 T to 8 T, reflecting updated safety evaluations for UHF MRI.[61] While these guidelines define acceptable exposure limits, the increased field strength nonetheless introduces additional practical safety considerations for both participants and operators. Furthermore, human studies at field strengths exceeding 7 T have so far not revealed any major or unexpected adverse effects; however, continued monitoring of participant tolerance and physiological responses remains essential to ensure safety as UHF MRI continues to advance.[62-64]

From a safety perspective, the main MRI risk is the strong attraction of ferromagnetic objects. Most materials, including human tissue, are magnetically active and generate a secondary field in the presence of an external magnetic field. Magnetic susceptibility (χ) describes how strongly a material is magnetised relative to the B_0.[45] Furthermore, limited information is available from vendors regarding the safety and MR compatibility of implanted devices at 7 T, necessitating enhanced safety precautions.[12,45] Compared with clinical 1.5 T and 3 T MRI, more extensive screening protocols are required. As many 7 T examinations are performed in research settings, participants may not receive a direct clinical benefit.[45]

The gradient system in an MRI scanner, a major source of acoustic noise and PNS, does not necessarily operate at higher amplitudes or slew rates than those of a 3 T system. However, it is often operated at higher performance levels than those typically used in conventional 3 T clinical MRI protocols. This is particularly relevant in high-resolution imaging and fMRI applications, where rapid gradient switching makes it a key consideration for safety at UHF.[65] Additionally, the effect of B_0 inhomogeneity also increases at higher field strengths, leading to more

pronounced susceptibility artefacts, particularly in echo planar imaging (EPI) sequences used in fMRI.

When increasing the magnetic field strength, the Larmor frequency rises proportionally, from approximately 128 MHz at 3 T to 298 MHz at 7 T.[66,67] This increase in frequency is associated with higher RF energy deposition in tissue. Consequently, the SAR, which scales approximately with the square of the magnetic field strength, also increases at UHF strengths.[45,54] Head SAR in MRI systems is typically limited to 3.2 W/kg in both normal operating mode and first level controlled operating mode according to IEC and U.S. Food and Drug Administration (FDA) safety guidelines.[61,68] At 7 T, head SAR is the primary safety constraint, as RF transmission is presently performed using local transmit coils because no RF body coil is currently integrated into the scanner bore.

Concerning the three basic electromagnetic fields used in MRI, the most obvious difference at UHF strengths is the increased static magnetic field. Exposure limits for both gradient fields and RF fields are generally the same at UHF as at lower field strengths, as they are defined by established safety standards rather than being directly scaled with field strength.[7,61,68]

Although safety considerations in 7 T MRI encompass a broad range of factors, the focus of the present thesis is primarily on acoustic noise and PNS as physiological effects associated with UHF MRI. Accordingly, the discussion of safety considerations will be limited to these aspects.

It should nevertheless be emphasised that clinical implementation of 7 T MRI requires a comprehensive safety perspective, in which multiple factors must be balanced within a risk–benefit framework. Such an approach is essential to optimise scan protocols and ensure both patient safety and comfort in UHF MRI applications.

Acoustic noise exposure

Acoustic noise, which consists of pressure waves transmitted through the air, can affect auditory health. Its intensity is expressed on a logarithmic scale in decibels (dB).[4] Hearing threshold, a key indicator of auditory health, declines with age, genetic factors, ototoxic drugs, and cumulative noise exposure. Temporary threshold shifts (TTS) reflect auditory fatigue, causing sounds to seem muffled after exposure to loud noise, although thresholds generally recover within a day.[34,69-71]

Exposure to brief sound pressure levels (SPLs) above 100 dB LAFmax (Maximum A-weighted sound level with fast time weighting) may lead to transient hearing loss, temporary threshold shifts, or disturbances in cochlear microcirculation.[45,72] These high noise levels highlight the need for effective hearing protection in MRI environments. The IEC recommends using protection when levels exceed 99 dB(A),[61] although such levels may still pose risks for individuals sensitive to certain

frequencies or intensities. Similarly, the ICNIRP advises providing hearing protection from 80 dB(A) and requires it for levels above 85 dB(A).[73] According to the Swedish Work Environment Authority, LAFmax SPL should not exceed 115 dB, and the maximum instantaneous C-weighted peak (LCpeak) should remain below 135 dB(C).[74]

The level of sound produced by an MRI scanner varies over time throughout the sequence, e.g., due to changes in amplitude of the phase-encoding gradient pulses. To measure acoustic noise over a time interval, the parameter A-weighted equivalent continuous level (LAeq) is used.[4] The total daily 8-hour exposure should not exceed an LAeq of 85 dB(A).[75,76] Together, these guidelines and regulations emphasise the importance of implementing adequate protective measures to safeguard auditory health during MRI procedures.

In UHF MRI systems, the increased static magnetic field strength not only enhances imaging capabilities but also intensifies acoustic challenges within the scanner environment. Consequently, as higher B_0 increases the Lorentz forces acting on the gradient coils, system operation results in stronger vibrations and elevated noise levels.[77,78] However, increased vibration of the gradient coils can also lead to greater mechanical damping, meaning that vibrations and acoustic output do not scale linearly with B_0, a phenomenon known as Lorentz damping.[77] In practice, improved acoustic and vibration insulation, together with the greater inertial mass of higher-field magnets, reduces the transmission of vibrations. This limits the increase in acoustic noise, resulting in only slightly higher SPLs at 7 T compared with 3 T. Conversely, the increased intrinsic SNRs at higher field strengths are often used to achieve higher spatial resolution, which requires stronger gradient amplitudes and can therefore increase acoustic noise.[45]

In addition, certain MRI pulse sequences, particularly three-dimensional (3D) acquisitions and techniques involving rapid gradient switching, such as EPI sequences used in fMRI and Diffusion Tensor Imaging (DTI), can generate LAFmax levels of 110–130 dB.[33,53,79,80] These types of sequences are widely used, especially in 7 T environments.[56,81-86]

Peripheral nerve stimulation (PNS)

There is no fixed regulatory upper limit for gradient amplitude (mT/m) in MRI systems. Instead, safety considerations are primarily based on physiological effects, such as PNS, as outlined in standards and recommendations from organisations including the IEC, the FDA, and the ICNIRP.[61,68,73]

PNS is more likely to occur during rapid imaging sequences with fast gradient switching and is therefore also expected to be frequent at 7 T (as described above). This is supported by previous studies reporting that more than 40% of participants experience PNS during 7 T examinations.[20,87]

To minimise PNS, MRI scanners estimate a predicted PNS level for each sequence. This estimation is expressed as a percentage, where 100% corresponds to a gradient output level at which approximately 50% of individuals are expected to experience PNS.[88]

Limits for minimising PNS in gradient systems are based on direct measurements from studies involving human volunteers. The established limits are defined such that, in Normal Operating Mode, the gradient system shall not exceed 80% of the PNS threshold, whereas in First Level Controlled Operating Mode, operation up to 100% of the PNS threshold is permitted.[61] First Level Controlled Operating Mode is frequently used in 7 T MRI settings.[45] In this mode, the upper dB/dt limit (T/s) can be expressed as 60000/(ramp time in μs). This limit is implemented to prevent cardiac stimulation, as described by Schaefer et al. (2000) and derived from IEC 60601-2-33 safety standards for gradient switching. It should be acknowledged that scanning performed within the First Level Controlled Operating Mode recommends medical supervision of the patient.[41,61]

FDA guidelines recommend that individual PNS studies involving at least 11 subjects should be performed when dB/dt exceeds 20 T/s, in order to determine thresholds for both mild and painful stimulation.[89] Although these regulatory guidelines apply to all MRI gradient systems, PNS assessment is particularly relevant for systems operating at higher gradient amplitudes and slew rates.[42] Previous research has shown that modifying gradient waveforms and gradient rise times may reduce PNS while maintaining imaging performance.[90,91]

The occurrence and perception of PNS are influenced not only by gradient performance, such as amplitude and slew rate, but also by individual anatomical and physiological characteristics. Simulation studies have demonstrated that body size, tissue composition, and fat distribution may affect the magnitude and distribution of induced electric fields. Adipose tissue may act as an electrical insulator between conductive muscle regions, potentially reducing the electric field experienced by nearby nerves and thereby increasing the threshold for stimulation.[92] Furthermore, evidence suggests that PNS is strongly dependent on participant positioning within the scanner.[93] Consequently, considerable interindividual variability exists in both objective PNS levels and subjective perception during MRI examinations.

Human response to the 7 T MRI environment

This section addresses both physiological responses and subjective experiences during 7 T MRI examinations. Physiological effects refer to objectively measurable changes in the body, such as alterations in cochlear outer hair cell function or PNS, whereas perceptual effects describe the individual's subjective experience, including noise perception, tingling sensations, and discomfort.

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Perceptual effects can be viewed as subjective manifestations of underlying biological processes, as physiological responses to the 7 T MRI environment may give rise to sensations such as dizziness, disorientation, headache, nausea, metallic taste, or visual phenomena.[20,23,87,94-96] In addition to auditory effects and PNS, potential cognitive effects have also been highlighted as a relevant aspect of exposure to the 7 T MRI environment; however, findings remain sparse and inconclusive.[65]

Experimental studies support a vestibular contribution, showing that exposure to static magnetic fields ≥7 T can induce circling behaviour in animal models. This is accompanied by increased c-Fos expression, a marker of neuronal activation also found after vestibular stimulation such as rotation.[97,98] Human studies on the visual system consistently demonstrate a significant reduction in near visual contrast sensitivity, suggesting a specific susceptibility of visual processing to static magnetic fields.[99,100] This indicates activation of brain regions involved in balance and spatial orientation, providing a plausible biological basis for perceptual effects such as dizziness and disorientation in humans.

Nevertheless, most cognitive domains have shown inconsistent or non-significant effects of UHF.[100-103] Overall, no clear relationship has been observed between magnetic field strength and the magnitude of cognitive effects, and the limited evidence does not allow firm conclusions regarding broader cognitive impacts. However, a common limitation across some of these studies is that cognitive assessments were typically conducted without concurrent MRI sequence acquisition. Since different electromagnetic field components may exert distinct physiological and perceptual effects, there may be value in considering the MRI examination as a whole experience. Accordingly, there is a need for studies that assess subjective and cognitive responses during MRI acquisition itself, to better characterise potential effects of UHF exposure under realistic scanning conditions.

The occurrence and intensity of the human response vary depending on factors such as field strength, temporal variation in electromagnetic fields, body position, and movement within the magnetic field. Furthermore, individual susceptibility influences both the likelihood of experiencing such effects and how they are perceived.[65] In addition to physiological and directly perception-related effects, the individual experience during 7 T MRI may also be influenced by psychological factors.[104]

Assessing perceptual effects is crucial for understanding how participants experience the MRI examination, ensuring person-centred care, and providing a

foundation for strategies to improve safety, comfort, and overall tolerance. By combining both objective physiological measurements and subjective reports, this work captures a comprehensive picture of individual responses to UHF MRI, which is essential for the optimisation of imaging protocols and participant management.[65]

Strategies to reduce adverse effects

Acoustic noise generated by some MRI systems may cause hearing damage if exposure levels are sufficiently high and prolonged.[69] In addition to the potential risk of auditory injury, MRI examinations may cause discomfort and disorientation due to acoustic noise and the confined space.[105] Radiographers have also identified noise as a common source of patient complaints, which in some cases may interfere with the completion of the examination, and in occasional cases, acute tinnitus has been reported.[29,106] Taken together, these findings highlight the importance of system-oriented and person-centred approaches that balance workflow demands with persons' well-being and safety.[107] Moreover, acoustic noise not only affects a person's experience but also induces unwanted physiological stimulation, which interferes with fMRI data interpretation.[37,52]

Strategies to manage acoustic noise in MRI include a combination of person-centred and technical approaches. These typically involve adequate participant preparation and information, provision of hearing protection, and the implementation of sequence optimisation techniques aimed at reducing acoustic output during image acquisition.[69,108]

Passive noise reduction

Passive noise reduction involves physical barriers or materials that reduce sound transmission. Examples include earplugs, earmuffs, sponge pads, foam mats in the gap between head and coil,[109-112] or sound-attenuating bore liners.[113] These reduce the intensity of sound reaching the participant’s ears but do not change the source of the noise.

Proper training in earplug use is essential to ensure effective hearing protection. Earplugs must be rolled, the pinna manipulated, and the plug inserted correctly, as improper folding or squeezing can prevent it from expanding symmetrically and reduce its protective effect. Insertion depth should be assessed visually.[69,114] Even a small adjustment in insertion, around 3 mm, can significantly increase protection.[115] Deep insertion may trigger normal sensitivity or a coughing reflex, but it does not pose a risk of eardrum damage. Foam earplugs should not be cut or trimmed, as this may impair their ability to expand uniformly and achieve complete occlusion of the ear canal.[69] Dual hearing protection provides approximately 5 dB of additional

attenuation beyond the higher of the earplugs or earmuffs, assuming both hearing protection devices are used correctly.[116]

In summary, proper earplug size and correct insertion are essential for effective hearing protection, and training in the use and maintenance of hearing protectors should be provided.[116] The manufacturer-reported Noise Reduction Rating (NRR) or Single Number Rating should not be interpreted as actual attenuation in practice, as these values reflect laboratory-based estimates under optimal conditions with correct fit.[115]

Acoustic noise reduction

Acoustic noise reduction (ANR) can be categorised into hardware- and sequence-based methods. A hardware-based ANR reduces MRI acoustic noise by suppressing vibrations and limiting sound propagation from the coils, for example, by enclosing them in a vacuum chamber. This approach lowers noise levels without affecting the performance of imaging sequences.[51,52] Nonetheless, these approaches often increase system complexity and cost, and may in some cases reduce patient bore size as well as effective gradient performance.[117]

Sequence-based ANR, which controls gradient magnetic field waveforms, has been developed and is currently implemented by vendors across various MRI systems.[117-120] In sequence-based approaches, which are the focus of this study, acoustic noise is reduced by optimising gradient activity and/or avoiding acoustic resonance frequencies.[117] These approaches reduce mechanical vibrations of the gradient coils by adjusting the current slew rate, amplitude, and waveform shape. However, because ANR modifies the gradient waveforms, there is a potential concern regarding its impact on image quality.[121]

Optimising acoustic noise reduction and participant comfort

Sequence parameters on a given MRI scanner can be adjusted to reduce gradient amplitude, dB/dt, or the number of gradient pulses per second, which in turn lowers acoustic noise (Table 1).[4,45,122] These adjustments can be applied both during protocol setup and throughout the scanning session to improve patient comfort and tolerance. Careful consideration of potential effects on image quality is essential, and a multidisciplinary approach helps ensure a balance between diagnostic quality, MR safety, and patient experience by improving physical comfort. While such technical adjustments are important, optimising patient comfort also requires a broader person-centred approach grounded in professionalism, empathy, and effective communication.

Table 1. Reducing acoustic noise in practice.

Parameter	Adjustment for ANR	Explanation
Slice thickness	↑	Thicker slices require weaker gradient switching, leading to less mechanical vibration.
Field of view	↑	With constant bandwidth, a larger FOV requires lower gradient amplitude, resulting in reduced acoustic noise.
Pixel size	↑	Larger pixels require less steep gradients and slower gradient switching.
Repetition time (TR)	↑	Reduces gradient duty cycle (less frequent switching).
Echo time (TE)	↑	Allows a reduction in receive bandwidth, resulting in lower noise.
Bandwidth	↓	Lower bandwidth reduces gradient strength and slew rate demands.
Number of slices	↓	Less gradient activity per TR period, reduction of overall noise.
b-value	↓	Lower b-values require weaker diffusion gradients, resulting in less acoustic noise.

In addition, the operator requires a solid understanding of human responses to the MRI environment to appropriately mitigate potential effects. For instance, head movement within the main magnetic field may alter magnetic flux in the semicircular canals of the inner ear, generating magnetohydrodynamic forces on the endolymph and triggering vestibular responses.[123,124] Therefore, advising participants to avoid rapid head movements near the scanner may help reduce dizziness and related discomfort.

Physical comfort can also be improved by adjusting the participant's position to individual needs, supported by padding where necessary. As an example, previous findings suggest that hand position significantly influences susceptibility to PNS. Clasping the hands over the abdomen appears to reduce the dB/dt threshold and may increase the likelihood of tingling or painful sensations, likely due to the formation of a lower-resistance current loop through the upper limbs. The immediate cessation of symptoms upon hand separation further supports the role of conductive pathways in the arms, consistent with ulnar nerve activation at the elbow.[41,125,126]

The scanning environment should also be adapted in terms of temperature and lighting to further enhance comfort. These measures should be complemented by effective communication and an individualised approach to care, which are central to ensuring overall subject comfort.[65] This is supported by previous research describing MRI as an experience that may feel like "being in another world," where the unfamiliar environment and isolation within the scanner can challenge participants' sense of self-control. Perceived threats to control, coping efforts, and the need for support are closely interrelated. Effective communication, individualised information, and a trusting dialogue with the radiographer are crucial

for participants' ability to manage fear, discomfort, and feelings of loss of control.[127,128]

While person-centred strategies are essential for improving physical and psychological comfort, it is equally important to ensure high-quality, high-performance MRI examinations. This is necessary to meet the requirements of both diagnostic and research applications.

Evaluation of MRI performance and image quality

Given the need to maintain both diagnostic and research quality, it is essential to systematically evaluate the impact of operational and sequence-based strategies on MRI performance and image quality. Image quality in MRI is a critical determinant of diagnostic value, as higher spatial resolution does not necessarily translate into improved interpretability. This is particularly relevant in UHF MRI, where increased signal is accompanied by greater susceptibility to inhomogeneities and artefacts.[54] Commonly used parameters to assess image quality include SNR, contrast-to-noise ratio (CNR), spatial resolution and the presence of artefacts.[119,129-135]

The SNR is the ratio of the amplitude of the signal received to the average amplitude of the noise.[27] In an MRI, the individual voxels that constitute the image contain a mixture of signal and noise. The ratio of signal intensity to the noise level is referred to as the SNR.[6] It is a fundamental parameter for quality evaluation in MRI, and a higher SNR can be traded for improvements in spatial resolution or reduced acquisition time. Accordingly, SNR is widely used as a primary metric and is included in quality assurance programmes such as those established by the American College of Radiology (ACR).[67]

SNR and CNR are key determinants of image quality. While SNR describes the ratio between signal and noise, CNR describes the signal difference between adjacent tissues relative to noise. CNR is probably the most critical factor affecting image quality, as it directly determines the ability of the human eye to distinguish areas of high signal from areas of low signal.[27] For that reason, it also plays an important role in protocol development and sequence evaluation.[6]

In neuroimaging, these measures are commonly assessed across anatomical regions such as cerebrospinal fluid (CSF), white matter, and grey matter, as these tissues provide key contrast interfaces for evaluating image quality and tissue differentiation.[119-121,131,133,135,136]

Image quality evaluation in MRI is inherently linked to the presence of artefacts, which occur to some degree in all MR images. Artefacts are features that do not accurately represent the underlying anatomy and arise from physical, technical, or

person-related factors. They may manifest as signal loss, distortions, or ghosting, and can degrade image interpretability or obscure pathology.[6] In UHF MRI, artefacts are more pronounced due to increased sensitivity to magnetic field inhomogeneities and tissue susceptibility differences.[55]

While the evaluation of image quality provides objective metrics for assessing the performance of MRI sequences and technical strategies, it is ultimately the radiographer who integrates these technical considerations with person-centred care, ensuring that both the imaging objectives and individual experiences are addressed in a multidisciplinary context.

The role of the radiographer in the 7 T MRI environment

Radiographers possess a distinct combination of technical expertise, clinical knowledge, and a person-centred perspective, grounded in their direct involvement in imaging procedures. Their role at the interface between technology and patient care provides unique insights into both the practical implementation of imaging protocols and the lived experiences of participants.[137]

Radiological practice is continuously evolving, driven by the implementation of new technologies, which are changing the way patient care is delivered,[138] and by steadily increasing patient expectations. Ensuring high-quality care and optimal outcomes depends on the development and dissemination of rigorous research. Core areas such as image quality, MR safety, and person-centred care remain central, while ongoing advancements require the adoption of both new and established research methodologies.[139,140]

MRI poses risks if safety measures are not followed, making radiographers' adherence to protocols essential for protecting participants and staff.[141] Grounded in evidence-based practice, where research informs clinical decision-making,[142,143] multidisciplinary collaboration is essential, and radiographers play a key role in advancing research and shaping the future of medical imaging practice.[138,139]

Importantly, the radiographer is an integral part of the multidisciplinary team, contributing practical and clinical insights into how MRI protocols affect participants in a rapidly evolving technological landscape.[144] Implementing and optimising 7 T MRI requires balancing safety, participant comfort, and image quality to ensure both tolerable and diagnostically valuable scans (Figure 1). By integrating MRI physics, safety considerations, person-centred care, and evidence-based practice, the radiographer supports strategies to reduce adverse effects while maintaining image quality.

In this thesis, the radiographer's role is highlighted as integral to investigating the physiological and perceptual effects of 7 T MRI. By systematically evaluating

cognitive performance, auditory function, perceived noise levels, and PNS alongside subjective and objective image quality measures, this work aims to integrate safety, comfort, and image quality into a cohesive framework. This work emphasises how radiographers' engagement in both technical and person-centred aspects supports methodological development and evidence-based optimisation of UHF MRI, thereby bridging the gap between technical advancement and human experience.

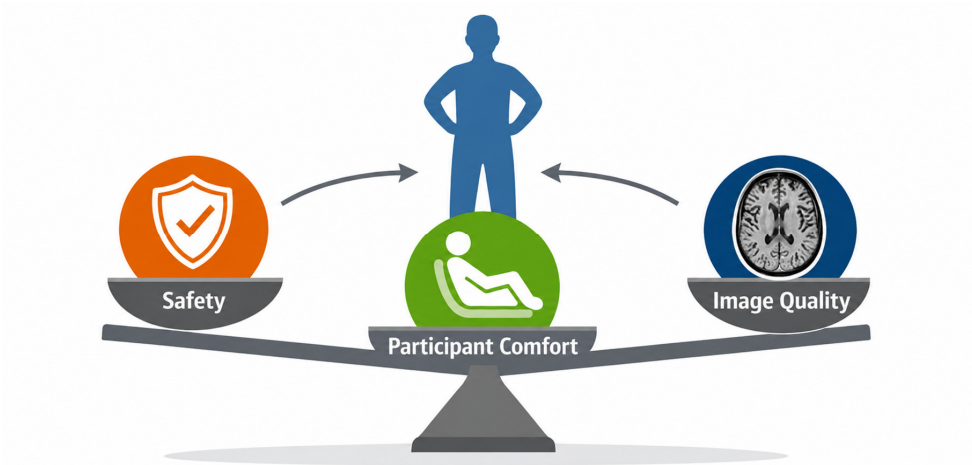


Figure 1. The radiographer´s role in the 7T MRI environment.
A conceptual illustration of the radiographer's central role in 7T MRI, depicted as the balancing point of a scale integrating key aspects: safety and care, advanced technical competence, and image quality optimisation. The figure highlights how the radiographer continuously balances these interconnected responsibilities to ensure safe, effective, and high-quality UHF MRI examinations. (Generated with assistance from OpenAI and adapted for illustrative purposes.)

Rationale

UHF MRI at 7 T provides advantages for neuroimaging due to its increased SNR and improved spatial resolution, and improved CNR, enabling enhanced visualisation of brain structures compared with lower field strengths. These capabilities have expanded the use of 7 T MRI in neuroscience research and show increasing potential for clinical applications. However, the transition to UHF MRI also introduces challenges related to the interaction between the MRI environment and the human body.

Exposure to the 7 T MRI environment may induce a range of physiological and perceptual effects. Previous studies have reported transient subjective sensations, such as dizziness, vertigo, and sensory disturbances, as well as possible effects on cognitive performance during exposure to strong static magnetic fields. Furthermore, the high sound pressure levels generated by rapidly switching gradient coils raise concerns regarding auditory effects, particularly during repeated exposure. PNS, caused by time-varying gradient fields, represents another important safety and comfort consideration in UHF MRI.

Although these phenomena have been investigated, important knowledge gaps remain regarding human responses to the 7 T MRI environment and how technological developments may mitigate these effects. Strategies aimed at reducing acoustic noise and gradient-related stimulation, therefore, require further evaluation.

Active noise reduction techniques, such as SofTone, have been introduced to reduce gradient-induced acoustic noise. However, evidence regarding their effectiveness in reducing acoustic exposure and perceived noise during UHF MRI remains limited, particularly concerning potential effects on PNS and image quality. Evaluations of such techniques should therefore include both physiological and perceptual outcomes, together with potential effects on image quality.

Consequently, a comprehensive assessment of both human responses to the 7 T MRI environment and technical strategies for mitigating adverse effects is needed. This includes investigating cognitive performance, auditory effects, and subjective experiences during 7 T MRI, as well as evaluating the effects of the SofTone noise reduction technique on acoustic noise, PNS, image quality, and participant comfort. Such investigations are important for addressing current knowledge gaps and supporting the safe and optimised implementation of 7 T brain MRI.

Aims

The overall aim of this thesis was to investigate physiological and perceptual effects associated with the 7 T MRI environment and to evaluate strategies for reducing adverse effects while assessing these strategies' effects on image quality in 7 T brain MRI.

Specific aims

I. To investigate the cognitive performance of healthy participants in a 7 Tesla MRI environment in connection with subjectively experienced effects.

II. To investigate the effects of repeated 7 Tesla MRI exposure on cochlear outer hair cell function in healthy adults, using a dual hearing protection method.

III. To evaluate image quality, sound pressure levels, and perceived noise levels when using the acoustic noise reduction technique SofTone and to compare with conventional imaging during 7 Tesla brain MRI.

IV. To investigate the effects of SofTone-based sequences during 7 Tesla brain MRI by evaluating peripheral nerve stimulation levels (PNS), perceived PNS, and image quality compared with conventional sequences.

Methods

Setting and recruitment

Data were collected at the Swedish National 7 T Facility, a collaborative initiative between Lund University and Region Skåne, located within the Centre for Imaging and Function at Skåne University Hospital in Lund.[145] The 7 T MRI system is accessible to researchers from all Swedish universities, with a primary focus on advanced MRI research and on fostering a multidisciplinary research environment.[11]

All participants were healthy volunteers, recruited specifically to undergo a 7 T MR examination in the context of a research study. Recruitment was conducted by the doctoral student via research portals, the study centre, the hospital and university networks. Eligibility was confirmed through verification of the inclusion and exclusion criteria before participation.

In **Paper I**, a control group was also included, consisting of participants who completed only the cognitive tests without undergoing the 7 T MRI examination. This group served as a comparison for the cohort performing cognitive testing in conjunction with the 7 T MRI procedure. Participants in the control group were approached regarding study participation during a scheduled teaching session at the University. Those who chose not to participate were instructed to leave the test materials blank, ensuring voluntary participation without any impact on their academic activities.

All MR examinations were conducted in First Level Controlled Operating Mode, not exceeding the SAR limit for the head of 3.2 W/kg, using an actively shielded 7 T MR scanner (Achieva, Philips, Best, The Netherlands). Each participant was examined head-first in a supine position.

The system is equipped with a gradient system featuring a maximum amplitude of 40 mT/m and a maximum slew rate of 200 mT/m/ms. The scanner has a tunnel diameter of 58 cm, a magnet length of 3.3 m, and a maximum spatial field gradient (dB/dz) of the stray field of 7.86 T/m at 130 cm from the isocenter. An 8 transmit (Tx)/32 receive (Rx) Nova head coil (Nova Medical, Wilmington, MA, USA) was used for brain examinations in **Papers I** and **II**, while a 2 Tx/32 Rx head coil was used in **Papers III** and **IV**.

For **Papers I** and **II**, data collection was conducted as part of a larger research project in conjunction with 7 T MRI examinations. The project investigated short-term second language vocabulary learning using virtual environments with varying levels of sensorimotor engagement and technological immersion, including assessments of cognitive performance and brain structural changes.[146]

Papers III and **IV** were conducted as part of a collaborative project with Umeå University, with the overall aim of investigating and gaining a deeper understanding of sequence-based ANR at UHF 7 T MRI. All data collection was carried out by members of the research project, primarily the doctoral students, me, Linda Wennberg and Anton Glans, together with the co-supervisor and principal investigator, Boel Hansson. All researchers involved were licensed radiographers with expertise in MRI.

All volunteers underwent routine MR safety screening in accordance with standard procedures at the national 7 T MRI facility in Lund, Sweden. A multi-step screening workflow was applied. This included verbal screening at recruitment and a standardised written questionnaire used for all MRI examinations at Skåne University Hospital. Upon arrival, participants changed into MR-safe clothing. This was followed by a verbal safety interview in the preparation room outside the scanner suite. Finally, participants passed through a ferromagnetic detector at the entrance to the 7 T scanner room. This redundant procedure was implemented to ensure participant safety by minimising the risk of undetected ferromagnetic objects or contraindications across multiple checkpoints.[44] In connection with the MRI examinations, additional data were collected using standardised assessment tools targeting cognition, subjective experiences, cochlear function, perceived noise levels, and PNS.

The four papers included in this thesis are summarised in Table 2.

Table 2. Overview of the four studies in this thesis

	Paper I	Paper II	Paper III	Paper IV
Objective	Cognitive performance in a 7 T MRI environment in relation to subjectively experienced effects	Impact of repeated 7 T MRI exposure on cochlear outer hair cell function using dual hearing protection.	Image quality, sound pressure levels, and perceived noise using SofTone compared with the conventional 7 T MRI sequence	Perceived PNS and image quality during SofTone sequences compared with conventional 7 T MRI imaging.
Design	Prospective within-subjects study	Prospective within-subjects longitudinal study	Prospective within-subjects study	Prospective within-subjects study
Number of participants	42 (exposed group) 40 (control group)	39	28	28
Data	Cognitive performance scores[a], Self-reported MRI-related sensory experiences[b]	Changes in DPOAE response levels across time points[c]	Peak SPLs in dBA[d], Self-reported perceived noise levels[e], Qualitative image ratings, Inter-rater metrics for image quality assessment[f]	Self-reported PNS sensations[g]. Quantitative image quality metrics (SNR and CNR) based on grey and white matter[h]
Data collection methods	Digit Symbol Substitution Test (DSST) [a], Computer-based questionnaire[b]	Repeated DPOAE testing[c]	Instrumental SPL recordings[d], Borg CR10 scale®[e], Qualitative image evaluation by standardised four-point ordinal rating scales[f]	Structured verbal questionnaire[g], Grey–white matter quantification using ROIs (SNR/CNR)[h]
Status	Published	Published	Published	In manuscript form

[a, b, c, d, e, f, g, h] Identical superscript letters appearing in both the data description row and the data collection method row indicate a direct correspondence between them. Specifically, the superscript letters identify which data were obtained using the respective data collection method.

MRI acquisition protocols

Participants who underwent cognitive testing before and after the MRI examination in **Paper I** completed the full MRI protocol as presented in Table 3. Cochlear function was also assessed before and after this MRI session to evaluate potential hearing effects, as part of **Paper II**.

Table 3. Acquisition parameters of the MRI sequences in Papers I and II

Sequence	TR (ms)	TE (ms)	Flip angle (degree)	Field Of View (mm^2/mm^3)	Matrix (mm)	SENSE (phase×slice)
Survey	11	4.93	15	250×250×50	112×116×3	no
Coil reference scan	8.0	0.49	1	705×470×705	48×31×47	no
SENSE reference scan	8.0	0.78	1	700×700×404	128×95×101	no
B_0 shimming	3.8	1.54	10	240×180×172	64×48×46	1×1
T1w 3-dimensional (3D) turbo field echo	8.0	2.7	7	212×242×190	304×345×543	1.4×1.4
MP2rage	6.8	2.4	5	224×224×180	320×320×257	2×1
T2w 3D turbo spin echo	267	3200	90	224×224×180	320×320×257	2×2.8
B_0 shimming map for rs-fMRI	1.54	3.8	10	240×180×172	64×48×48	no
rs- fMRI*	25	1800	55	224×232×149	112×116v68	3
Flow mapping× 4**	4.8	15	7	200×220	400×440	2
DTI anterior-posterior direction	65	8010	90	225×224×160	112×109×80	3
DTI posterior-anterior direction	64	8437	90	225×224×160	112×109×80	3
FLAIR 3D	285	6000	55	230×230×168	192×190×140	2×1
DREAM 25° ***	2.4	0.99	12	240×180×240	64×48×64	no
DREAM 40° ***	2.4	0.99	12	240×180×240	64×48×64	no
DREAM 60° ***	2.4	0.99	12	240×180×240	64×48×64	no

*Resting state (rs) fMRI with MultiBand (factor 2), 35 dynamic scans with a dynamic scan time of 1000 ms and an Echo planar imaging (EPI) factor of 39.

**Velocity encoding (Venc) of 100 cm/s of the internal carotid artery and the vertebral artery

***Dual Refocusing Echo Acquisition Mode-a multislice B_1-mapping method obtained at different angles, where the angle denotes the STEAM preparation flip angle

In addition, participants took part in a language learning intervention and returned to the MRI facility after a brief learning session on the same day to undergo a second MRI scan with a slightly reduced protocol, in which the flow mapping and FLAIR sequences were omitted. The first MRI session had a total sequence duration of 50 minutes and 42 seconds, whereas the second MRI session lasted 39 minutes and 54 seconds.

For **Papers III** and **IV**, the MRI protocol was designed to evaluate the performance of the software-based ANR, SofTone. The imaging protocol consisted of two commonly used morphological sequences at the study centre: a T_2-weighted fast spin echo (T_2W FSE) sequence and a three-dimensional T_1-weighted turbo field echo (3D T_1W TFE). Each sequence was acquired twice, once using the

conventional (non-ANR) configuration and once using the SofTone-optimised ANR implementation (Table 4).

SofTone can be applied in five incremental steps (levels 1–5), each introducing progressively stronger gradient constraints to achieve additional acoustic noise reduction.[79] To maintain identical total acquisition times between the ANR and non-ANR conditions, minor sequence adjustments were required. For the sequence using SofTone, a slight increase in TE and an increase in receiver bandwidth were implemented, resulting in a modest reduction in SNR and CNR. For the T_2-weighted sequence, echo spacing was increased from 12 ms to 13.8 ms when SofTone was applied.

The protocol also included standard preparatory scans: a survey scan (min:sec; 00:57), a coil reference scan (00:11), a sensitivity encoding (SENSE) calibration scan (00:38), and a B_0 shimming sequence (00:18). In addition, DTI was acquired with and without SofTone; however, analysis of the DTI data falls outside the scope of this thesis.

All sequences were performed in a randomised order across participants, and participants were blinded to the applied sequence type to minimise potential bias.

Table 4. Acquisition parameters for the T_2W FSE and 3D T_1W TFE sequences in Papers III and IV

	T_2W FSE		3D T_1W TFE	
IMAGE PARAMETERS	Conventional	SofTone	Conventional	SofTone
Slice thickness (mm)	1.0	1.0	0.7	0.7
Matrix (mm^2)	440×366	440×366	304×343	304×343
Field of view (mm^2)	220×190	220×190	240×210	240×210
Acquired voxel size (mm^3)	0.50×0.52×1.0	0.50×0.52×1.0	0.70×0.71×0.70	0.70×0.71×0.70
Reconstructed voxel size (mm^3)	0.29×0.29×1.0	0.29×0.29×1.0	0.36×0.36×0.35	0.36×0.36×0.35
Slice orientation	Transversal	Transversal	Sagittal	Sagittal
Slice	48	48	543	543
NSA	2	2	1	1
Acceleration factor (SENSE)	1.7	1.7	1.4×1.4	1.4×1.4
Turbo factor	9	9	250	250
Scan time (min:sec)	08:26	08:26	08:40	08:40
SofTone factor	–	3	–	4
Bandwidth (Hertz/pixel)	125.9	142.6	252.5	336.6
TR (ms)	5060	5060	2.7	3.8
TE (ms)	60	69	8.0	8.0
Inversion time (ms)	–	–	1200	1200
Flip angle (°)	90	90	7	7
Echo spacing (ms)	12.0	13.8	–	–
Shot duration/interval (ms)	–	–	2210/3500	2210/3500

Data collection procedures and assessment tools

Data were collected using a multimodal approach combining subjective, physiological, and imaging-based assessments. Standardised protocols were applied across all measurement domains, and both qualitative and quantitative methods were used to evaluate participant responses and MRI performance.

Digit Symbol Substitution Test

In **Paper I**, we used a Digit Symbol Substitution test (DSST) to evaluate cognitive effects after the participants underwent a 7 T MR scanning session using a 50-minute-long research MRI protocol (Table 3). Participants first completed the paper-based DSST, followed by an MRI scan. After the scan, the DSST was repeated (Figure 2). All participants received verbal instructions about the test procedure and were shown an example of the DSST before testing to familiarise themselves with the task.

The DSST was a paper-and-pencil cognitive test presented on a single sheet, in which participants match symbols to numbers (1–9) according to a reference key at the top of the page (Figure 3). Symbols are copied into spaces below rows of randomly arranged numbers. The task is to correctly substitute as many symbols as possible within 120 seconds, proceeding row by row from left to right.[147,148] The test is scored by counting the number of correct symbols inserted.[149]

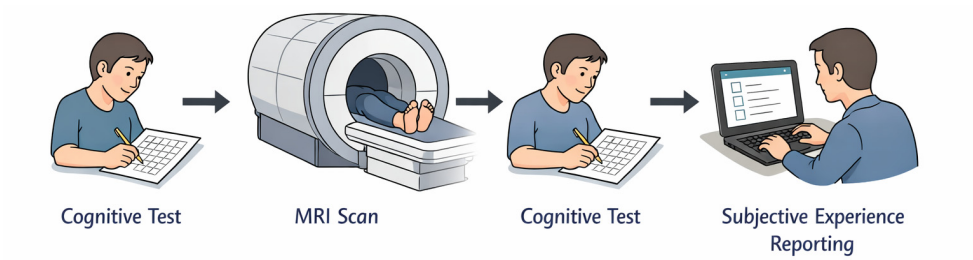


Figure 2. Flowchart of the data collection process for Paper I.
Schematic illustration of the study procedure in Paper I, including the DSST assessment and participants' reported subjective experiences in connection with a 7 T MRI scanning session. Note: Figure generated using AI (OpenAI) and adapted for illustrative purposes.

The strategy to solve the DSST is through repeated matching of numbers to their corresponding symbols using the reference key. The task involves temporarily encoding each number in short-term memory, visually scanning the key to identify the correct symbol, and retrieving the relevant information to guide the response. This process requires continuous alternation between the test sheet and the key, relying on visual search, working memory, and attentional control. Motor output is

integrated with these cognitive processes when the identified symbol is written down. The ability to filter out irrelevant information, such as visually similar symbols, also contributes to performance. This test has been shown to have high test–retest reliability.[148,150]

To reduce test–retest effects, alternate versions of the DSST were administered during the post-test following the MRI scan in **Paper I**. This involved modifying the reference key by assigning previously unused symbols to each number. The order of test versions was randomised and counterbalanced across participants.[151] Performance was defined as the number of correctly completed symbols within 120 seconds.

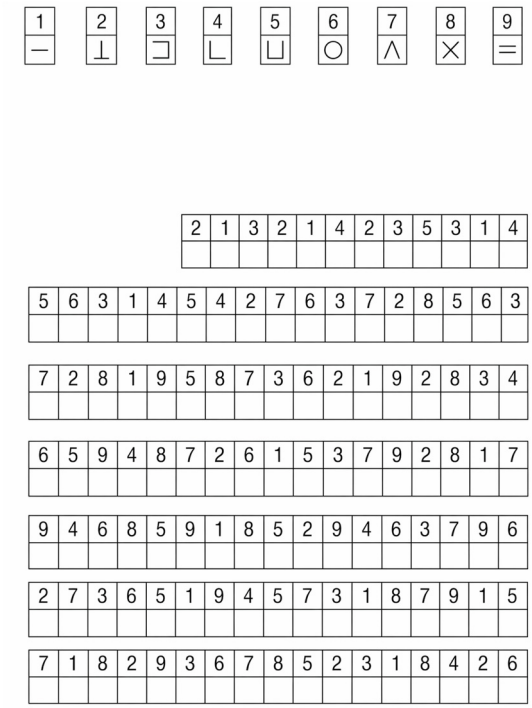


Figure 3. Digit Symbol Substitution test
DSST form with a reference key at the top and empty grids below, where participants fill in the correct symbols corresponding to each number.

In neuropsychology, the DSST is one of the most widely used tests due to its brevity, high reliability, and minimal susceptibility to influences from language, culture, or educational background. It is considered a sensitive measure of global cognitive functioning, with performance reflecting processing speed as well as contributions from attention, working memory, visuoperceptual abilities, and motor speed. As a paper-and-pencil test, it also requires fine motor coordination and visuomotor integration.[147,151]

Subjective experiences

Participants' subjective experiences were assessed in **Paper I** using a structured, computer-based questionnaire (REDCap; Research Electronic Data Capture) administered immediately after the MRI scanning session and subsequent cognitive testing (Figure 2). The questionnaire included previously reported short-term sensations associated with UHF MRI, such as dizziness, perceived motion, nausea, and headache, selected to capture relevant and commonly reported perceptual effects in this environment.[18,21,23,87,152]

Each sensation was rated using a 6-point Likert-type ordinal scale inspired by visual analogue scaling principles, ranging from "none" to "very much" (none, very little, little, moderate, much, very much).[23,44] This approach was chosen to combine the simplicity and clinical feasibility of categorical response options with the conceptual framework of intensity scaling derived from visual analogue methodology. Higher scores indicated stronger perceived sensations, and responses were treated as ordinal data in subsequent analyses.[153]

Subjective experience was also assessed in **Paper IV** regarding the presence and sensations of PNS in connection with conventional and ANR sequences. Immediately after completion of each imaging sequence, participants were asked verbally, while remaining positioned inside the scanner bore, whether they had experienced any twitching in any part of the body during image acquisition. The questions were also displayed on a screen visible to the participant during scanning. Responses were given verbally by the participants and recorded by the doctoral students involved in the project. If twitching was reported, follow-up questions were administered regarding the perceived intensity, any associated discomfort, and the specific body region(s) where the sensations were felt (Table 5). "Twitching" was defined for participants as the experience of unusual and/or involuntary muscle contractions. This definition was used to capture sensations potentially induced by rapidly switching gradient fields, corresponding to PNS.[20]

Table 5. Questionnaire items used to evaluate the occurrence and perception of PNS

Question	Response options
Did you feel any twitching in any part of your body during image acquisition?	None at all; Very slight; Slight; Moderate; Strong; Very strong
If you felt twitching in any part of your body during image acquisition, how intense was the twitching at its strongest?	None at all; Very slight; Slight; Moderate; Strong; Very strong
If you experienced twitching, how did you perceive it?	Did not experience any twitching; Not at all unpleasant; Very slightly unpleasant; Slightly unpleasant; Moderately unpleasant; Very unpleasant; Extremely unpleasant
Indicate the body part with the most pronounced twitching.	Head; Eyes; Neck; Back; Upper body; Abdomen; Pelvis; Hand; Arm; Leg; Foot

Borg CR10 scale

To evaluate how participants perceived the noise levels during the pulse sequences in **Paper III**, the Borg CR10 scale® was used.[154] The Borg CR10 scale is a category-ratio scale designed to measure perceived intensity, ranging from "Nothing at all" (0) to "Extremely strong / Maximal" (10). In exceptional cases, ratings above 10 can be used to indicate "Absolute maximum" intensity.[155]

The fixed upper limit of the scale is important for reflecting the concept of maximal intensity, allowing individuals to relate their perceived sensation to their own previous experiences. As such, the perceived maximum may vary depending on an individual's personal history, making the scale adaptable across different individuals.[156] The scale is also continuous, and the use of decimal values is permitted to allow more precise ratings.[155,156]

The Borg CR10 scale has demonstrated good reliability and validity across a wide range of applications, including the assessment of different types of perceived exertion and sensations, such as breathlessness, breathing difficulty, fatigue, and pain, particularly in the contexts of exercise testing, training, and rehabilitation.[157-159]

Because the Borg CR10 scale is a general intensity scale applicable to a wide range of perceived sensations and experiences, it has also been used to quantify subjective perceptions of sound intensity, including perceived noise levels during MRI examinations.[119]

In **Paper III,** participants were instructed to rate their perceived noise level after each pulse sequence using the Borg CR10 scale.[154] Standardised instructions on how to use the scale were provided in the preparation room immediately before the MRI examination. Participants were informed that they should estimate the intensity of the noise they experienced during each sequence. During scanning, the Borg CR10 scale was displayed on a screen visible to the participants while lying in the MRI scanner. After each sequence, participants verbally reported their rating through the MRI communication system. Decimal values were allowed to enable more precise reporting. The reported value represented the participants' perceived noise intensity for the respective pulse sequence.

Acoustic noise assessment

According to National Electrical Manufacturers Association (NEMA) guidelines, MR acoustic noise is characterised by repetitive acoustic impulses, requiring both LCpeak and LAeq for a complete assessment. LCpeak reflects instantaneous maximum sound pressure from individual gradient pulses and is relevant for acute auditory risk, while LAeq represents the time-integrated exposure and overall acoustic energy during scanning.[160]

In **Paper II**, SPLs for each MRI sequence were measured using a calibrated dosimeter (Svantek SV 104A) meeting IEC 61672-1 Class 2 specifications, with a measurement uncertainty of approximately ±1–2 dB.[161] For safety reasons, the dosimeter microphone was positioned in the MR control room, with one end of a rubber tube attached to the microphone and the open end placed inside the head coil at the isocenter of the MRI bore within the image volume (Figure 4).[160]

Calibration measurements were performed in advance to determine correction factors accounting for tube resonance and attenuation. The tubing was suspended to avoid contact with the MRI table or other surfaces, thereby minimising vibration transmission during scanning.

All acoustic measurements were performed by a licensed audiologist to ensure methodological reliability. Sound exposure was recorded for each MRI sequence as LAeq, LAFmax level over any 125 ms interval, and LCpeak.

Figure 4. Sound level assessment at the 7 T facility
Experimental setup for acoustic measurements. A rubber tube was suspended from the ceiling inside the MR room and routed into the MRI bore without contacting the patient table or surrounding surfaces. The open end of the tube was positioned at the isocenter inside the head coil. Sound levels were measured for each sequence in the imaging protocol used in Paper II.

In **Paper III,** Instantaneous peak SPLs were measured with an MRI-compatible sound level meter (OptiSLM 100; Optoacoustics Ltd, Mazor, Israel). For each participant, a microphone (OPTIMIC 1155) was placed at the jugular notch,[154] outside the head coil, due to space constraints and to minimise acoustic artefacts associated with placement within the coil. This positioning was chosen to maintain a consistent microphone location between sequences and participants, as noise levels vary along the bore.[162] Peak SPLs were recorded,[154] which indicate potential immediate hearing loss above a critical threshold.[69]

Distortion product otoacoustic emissions

Distortion product otoacoustic emissions (DPOAEs) were used to assess cochlear outer hair cell function in participants before and after 7 T MRI exposure in **Paper II**.[163] DPOAEs provide an objective measure of cochlear function and have previously been used in studies to assess potential effects of MRI-related acoustic noise on hearing.[110,111,164,165] This method allows for the detection of subtle changes in cochlear function[166-168] that may occur as a result of acoustic noise generated during MRI.

Otoacoustic emissions (OAEs) are sounds originating from the cochlea that can be detected using a microphone inserted into the ear canal. These emissions result from the mechanical vibrations of the sensory hair cells in the cochlea as they energetically respond to auditory stimuli. One method of OAE recording, DPOAEs, involves playing two different tones into the ear simultaneously. Inside the cochlea, these two tones interact in a way that produces additional sounds.[166,169] These additional sounds, which are not part of the original tones, can then travel through the ear canal and be detected by the microphone.

DPOAE measurements were obtained in a sound-attenuated room adjacent to the MRI facility, isolated from scanner and machine-room noise. Measurements were performed using an Interacoustics Titan system with the DPOAE module (firmware version 1.13.23, Titan Suite 3.6.0). Primary tone levels were set to L1 = 65 dB SPL and L2 = 55 dB SPL, with an f2/f1 ratio of 1.22. Before DPOAE testing, a visual otoscopic examination was performed using an otoscope to ensure that the external auditory canal was free from obstructing debris and that the tympanic membrane was intact.

Participants were first tested with DPOAE prior to the MRI examination to obtain baseline cochlear function. Immediately after completion of the initial MRI session, DPOAE measurements were repeated to assess any acute changes. The participants underwent an additional MRI session on the same day, followed by a second post-MRI DPOAE assessment. Finally, participants returned on a separate day for another follow-up DPOAE measurements to evaluate potential longer-term effects of MRI exposure (Figure 5). All measurements were conducted under standardised conditions to ensure reproducibility and reliability.

DPOAE amplitudes were measured in both the left and right ears at four time points across seven frequency bands ranging from 1 to 6 kHz. This approach allowed for the detection of both immediate and delayed changes in cochlear outer hair cell function related to exposure to the 7 T MRI environment.

Figure 5. Flowchart of the longitudinal data collection process in Paper II.
The flowchart illustrates the study timeline, including baseline measurements, two MRI examinations, each followed by DPOAE measurements, and a final follow-up visit with DPOAE assessments performed on a separate day.

Assessment of image quality

To evaluate the ANR technique SofTone to improve our understanding of sequence-based ANR at UHF (7 T) MRI, we used a qualitative image assessment in **Paper III**[154] and a quantitative image assessment in **Paper IV**. The rationale for employing both qualitative and quantitative image quality assessment in this work was as follows: Qualitative assessment may provide clinically relevant information regarding anatomical detail, artefacts, and overall diagnostic usability of MRI data. However, such assessments are inherently subjective and may be influenced by inter-observer variability.[129,170] Quantitative analysis was therefore included as a complement to the qualitative assessment, providing objective and reproducible measures such as SNR and CNR. By combining qualitative and quantitative approaches, the evaluation was intended to improve both clinical relevance and methodological robustness, thereby enabling a more comprehensive comparison between the modified and standard imaging sequences.

Qualitative image assessment

Qualitative image assessment in **Paper III** was carried out by three neuroradiologists (readers), each with more than 10 years of experience in neuroradiology. The assessed sequences comprised T_2W FSE and 3D T_1W TFE, acquired both with and without ARN.

All images were evaluated independently, and the readers were blinded to subject-related information as well as to the imaging sequence. To minimise bias, the order of image presentation was randomised for each subject, and the sequences were presented in a blinded manner. Scoring was performed independently across readers and across the randomised image sets.

The qualitative image assessment was inspired by previous publications on MRI sequence evaluation using a 4-grade scale.[119,171] It included evaluation of the occurrence and distribution of artefacts, the type of artefacts and their impact on image interpretation. In addition, overall image quality, the ability to differentiate between white- and grey matter, and the distinction between cerebrospinal fluid (CSF) and surrounding soft tissues were evaluated (Table 6).[32,131-133]

Table 6. Qualitative MRI image assessment criteria

Domain	Score scale	Definition
Artefacts (impact)	1-4	1 = Non-diagnostic image due to severe artefacts 2 = Major artefacts significantly interfering with interpretation 3 = Minor artefacts with limited impact on interpretation 4 = No artefacts affecting diagnostic interpretation
Artefact type (if present)	Categorical	Motion / Susceptibility / Flow / Signal loss / Other (Specify)
Overall image quality	1-4	1 = Non-diagnostic image quality 2 = Limited diagnostic quality with clear limitations 3 = Good diagnostic quality with preserved anatomical detail 4 = Excellent image quality with no diagnostic limitations
White- and grey matter differentiation	1-4	1 = No discernible separation 2 = Poor separation with limited structural definition 3 = Good separation with clear anatomical delineation 4 = Excellent separation - sharp contrast between WM and GM
CSF–soft tissue differentiation	1-4	1 = No discernible separation 2 = Poor separation with limited contrast 3 = Good separation with clear delineation 4 = Excellent separation - strong contrast and clear boundaries

Quantitative image assessment

SNR and CNR were used as quantitative metrics for MRI image quality assessment in **Paper IV**. These were applied to compare imaging sequences with and without SofTone. While SNR reflected the overall signal relative to noise, CNR quantified differences in signal intensity between anatomical regions and was therefore more directly related to tissue differentiation.[172]

In all 3D T_1W TFE sequences, acquired both with and without SofTone, regions of interest (ROIs) were manually delineated in grey and white matter to enable quantitative assessment of SNR and CNR. ROI placement was performed using Horos® (Version 4.0.1, Annapolis, MD, USA). All ROIs were defined with an identical size (5.7 mm²) to ensure comparability across measurements. Grey matter ROIs were oval-shaped and placed in the cingulate gyrus cortex to maximise cortical coverage, while white matter ROIs were circular and positioned in the frontal part of the corpus callosum. A predefined ROI template was applied to each image sequence, and positioning was adjusted when necessary to account for minor subject motion between acquisitions.

SNR was calculated using Equation [1], where $\mu_{foreground}$ is the mean signal intensity, and $\sigma_{foreground}$ is the standard deviation within the foreground mask, defined as the combined white matter (*WM*) and grey matter (*GM*) regions. The parameter *n* denotes the number of voxels within the mask. A correction factor $\sqrt{(n/(n-1))}$ was applied to account for sample size bias.[132,135]

CNR was calculated using Equation [2], where μ_{VM} and μ_{GM} represent the mean signal intensities of white- and grey matter, respectively, σ_{WM} and σ_{GM} represent

their corresponding standard deviations. CNR was defined as the absolute difference in mean signal between white matter and grey matter, normalised by the combined noise term $\sqrt{(\sigma_{WM}^2 + \sigma_{GM}^2)}$, with the same $\sqrt{(n/(n-1))}$ correction applied.[132,135]

$$\text{SNR} = \frac{u_{foreground}}{\sigma_{foreground}} \cdot \sqrt{\frac{n}{n-1}} \quad [1]$$

$$\text{CNR} = \frac{abs|\mu_{WM} - \mu_{GM}|}{\sqrt{\sigma_{WM}^2 + \sigma_{GM}^2}} \sqrt{\frac{n}{n-1}} \quad [2]$$

Data management and statistical analyses

Within all projects included in this thesis, quantitative statistical methods have been applied for data management and analysis. Both parametric and non-parametric approaches were used, depending on the characteristics and distribution of the data, as well as the specific research questions addressed. An overview of the statistical methods applied in the different studies is provided in Table 7.

Data from the studies included in this thesis were analysed using SPSS software (IBM Corp., Armonk, NY, USA; versions 25 or 27). In **Papers I** and **II**, data were also analysed using JASP, an open-source statistical software program.[173]

Table 7. Statistical analyses used in the individual papers.

	Paper I	Paper II	Paper III	Paper IV
Descriptive statistics	✓	✓	✓	✓
Paired- samples t-test	✓	✓	✓	✓
Independent t-test	✓			
Repeated measures analysis of variance	✓			
Spearman´s correlation test	✓			
Bonferroni adjustment for multiple comparisons		✓		
Wilcoxon´s signed rank test			✓	✓
Krippendorff's alpha			✓	

In all studies involving paired data, parametric analyses were applied when the data were normally distributed. This included paired t-tests for within-subject comparisons. In **Paper I**, a paired t-test was used to compare cognitive performance on the DSST before and after the MRI examination.

In **Paper II**, paired t-tests were used to compare DPOAE measurements obtained at different time points in relation to the MRI examinations. As measurements were collected at seven frequencies on both ears, Bonferroni correction was applied to adjust for multiple comparisons, resulting in a significance threshold of $p < 0.0036$. Given the repeated-measures design, different analytical approaches were initially considered. In consultation with a statistician, paired t-tests were selected for the primary analyses comparing baseline with each follow-up time point. This approach was chosen as the primary aim was to assess within-subject changes relative to baseline, without adjustment for confounding variables. Although mixed-effects models were considered, they were not expected to materially alter the results in this context, as the same predefined comparisons would still be evaluated. Paired t-tests were therefore considered appropriate given the study design and analytical objectives.

In **Paper III**, paired t-tests were used to compare SPLs and perceived noise levels (using Borg CR10 scores) between MRI sequences with and without SofTone, when data were normally distributed. When normality assumptions were not met, the Wilcoxon signed-rank test was used. The same non-parametric approach was used for qualitative image quality ratings, comparing sequences with and without SofTone.

In **Paper IV**, the Wilcoxon signed-rank test was used to compare scanner-derived PNS levels, as the data were not normally distributed according to the Shapiro–Wilk test. Perceived PNS ratings, which were ordinal in nature, were also analysed using Wilcoxon signed-rank tests. For quantitative image quality assessments, paired t-tests were used to compare conventional and noise-reduced sequences in terms of SNR and CNR.

Since **Papers I** and **IV** focused on individual subjective sensations rather than a composite construct, each item was analysed separately using ordinal statistical methods. Likert-type items were analysed using descriptive statistics and non-parametric methods, including frequencies, medians, and modes, as the data are ordinal and response categories cannot be assumed to have equal intervals.[174,175]

Ethical considerations

All studies included in this thesis were conducted in accordance with the Declaration of Helsinki[176] and were approved by the ethics committee. For **Papers I** and **II**, ethical approval was obtained for the cognitive testing and DPOAE measurements (DNR 2019-05387), as well as for subjective self-report questionnaires (DNR 2020-06907). For the joint research project with Umeå University, encompassing **Papers III** and **IV**, ethical approval was granted under a separate application (DNR 2022-05816-01).

All participants received both written and verbal information about the study. Participants were informed that they could withdraw from the study at any time without providing a reason. The studies were conducted in healthy volunteers to avoid exposing vulnerable patient populations to experimental procedures. All investigations were non-invasive and posed minimal risk to participants.

To ensure participants' safety, everyone undergoing MRI scans needed to complete a safety evaluation checklist to exclude any safety risks.[2,141] This was carried out in adherence to clinical MR safety standards, using the multi-step screening procedure described in the Setting and Recruitment section of this thesis.[44]

Established routines for acoustic noise attenuation in 7 T MRI were applied to all participants. Each participant was provided with double hearing protection, consisting of earplugs in combination with sound-absorbing impression paste. Verbal communication was maintained at regular intervals throughout the MRI scanning session to ensure participant well-being. In addition, all participants were equipped with an alarm device, enabling immediate contact with the MR radiographer if needed during the scanning session.

Morphological imaging was included in all examinations and was reviewed by a neuroradiologist for potential incidental findings. Any clinically relevant findings were reported to the appropriate clinical pathway for further follow-up.

The ethical framework of the work was guided by the professional ethical code for radiographers, emphasising respect for autonomy, patient safety, and minimisation of harm.[177] Furthermore, the study adhered to the European Federation of Radiographer Societies (EFRS) Code of Ethics, which underscores the responsibility of radiographers to ensure safe, evidence-based practice and the protection of research participants throughout the imaging process.[178]

Results and Discussion

The present findings can be understood within a conceptual framework integrating (1) participant experience, (2) physiological and perceptual responses, and (3) technical MRI performance, which together influence outcomes in UHF MRI. Within this framework, effective implementation of strategies aimed at balancing safety, comfort, and image quality requires interprofessional collaboration.

Radiographers play a vital role in this process, as their involvement in MRI examinations provides practical insight into all three domains. This positions them at the interface between participant-related factors and technical MRI performance, enabling the integration of these perspectives in clinical practice.

Participants may experience effects such as acoustic discomfort, PNS, and altered perceptions during scanning, underlining the importance of considering the human impact of the UHF environment alongside technical performance measures. The results further demonstrate that technical strategies, including passive and active noise reduction, sequence optimisation, and gradient management, play a central role in mitigating adverse effects while maintaining diagnostic image quality. However, these interventions involve trade-offs between reducing acoustic noise and PNS and preserving imaging performance.

Cognitive performance and subjective effects

Participants completed cognitive testing immediately before and after the 7 T MRI examination in **Paper I**. Following the MRI session, the time interval between exiting the scanner bore and initiating the post-test ranged from 36 to 180 seconds. No statistically significant differences were observed in the paired comparisons of cognitive performance before and after the MRI examination. In contrast, the control group, which completed the cognitive tests before and after a lesson, demonstrated a statistically significant improvement in the paired comparison (Table 8).

Both groups showed improved performance at post-test. Overall, the MRI group achieved higher scores than the control group at both pre-test and post-test; however, the magnitude of improvement was greater in the control group.

Table 8. Cognitive test performance in the MRI and control group.

Data are presented as mean ± SD at pre- and post-assessment. Within-group changes (Δ mean ± SD) and *p*-values from paired comparisons are reported.

Group	Test	Score (mean ± SD)	Score difference (Δmean ± SD)	*p*-value
MR group (n=42)	Pre-test	74.40 ± 15.76		
	Post- test	74.79 ± 15.37	-0.38 ± 7.41	0.74
Control group (n=40)	Pre-test	58.83 ± 11.88		
	Post- test	64.90 ± 12.68	-6.08 ± 5.90	<0.001

Of the 42 participants in the MRI group, 39 completed the questionnaire on subjective effects, and 37 reported experiencing effects to some extent (Figure 6). In total, 34 participants (87%) reported experiencing dizziness, 19 (49%) reported sensations of unrealistic movement in connection with passage through the scanner bore, 15 (38%) reported nausea, and 13 (33%) reported headache, with these effects generally being reported as mild in severity.

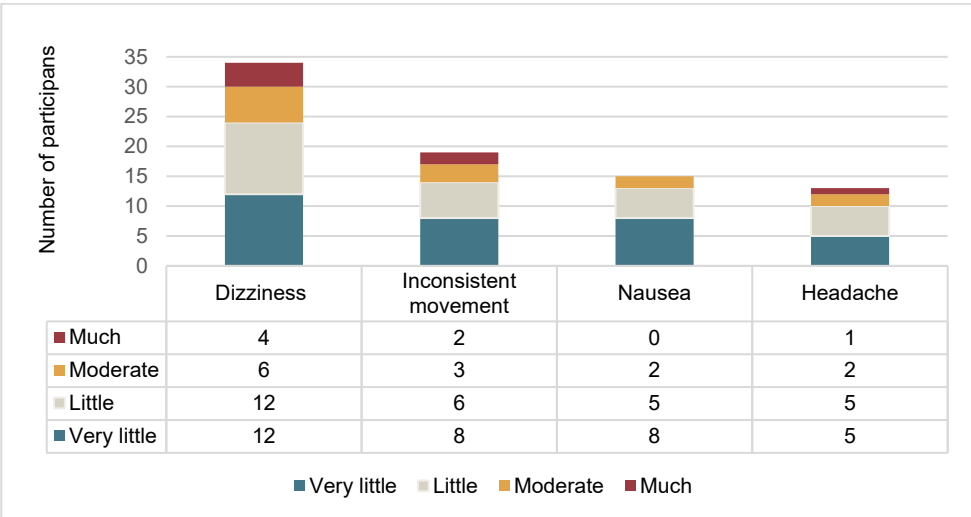


Figure 6. Self-reported MRI-related subjective effects.

The figure shows the distribution of subjective effects across symptom categories after MRI examination. Responses are presented as the number of participants reporting each severity level, ranging from no symptoms to more severe experiences. Responses *"none"* are excluded from this figure for clarity.

Although most participants reported subjective effects to varying degrees, we did not observe any statistically significant correlations between changes in cognitive test scores and self-reported symptoms in the MRI group. This includes dizziness

(Spearman´s rank $r_s(39) = 0.068$, $p = 0.680$), inconsistent movement ($r_s(39) = 0.065$, $p = 0.695$), nausea; ($r_s(39) = 0.263$), $p = 0.106$, and headache ($r_s(39) = 0.045$, $p = 0.784$).

In line with the findings in **Paper I**, previous studies investigating exposure to UHF MRI have generally reported no significant effects on cognitive performance in healthy adults. Exposure to magnetic fields up to 7–9.4 T, including conditions such as 9.4 T static field exposure and multinuclear MRI, has not been shown to significantly affect vital signs or cognitive function.[179,180] Similarly, studies at field strengths up to 8 T have concluded that no short-term cognitive effects are observed following exposure, and that cognitive performance remains largely unchanged.[100,103] Even when isolated statistically significant findings have been reported, such as in specific memory tasks, the effects have been extremely small and not considered clinically meaningful.[102,124]

However, a key limitation of these studies is that many have investigated the effects of static magnetic fields in isolation, without the context of a full MRI examination. This leaves an important gap in the literature regarding how the complete MRI environment, comprising static magnetic fields, gradient switching, acoustic noise, and the scanning context itself, may influence cognitive performance from a more holistic perspective. Consequently, further research is needed to better understand cognitive responses under realistic UHF MRI conditions.[99]

A previous study, together with our **Paper I**, has reported improved performance in post-MRI cognitive testing following scanning sessions. However, such improvements are most plausibly explained by learning effects rather than any true effect of UHF MRI exposure.[181] Overall, the literature indicates that static magnetic fields up to 11.7 T, the largest field available for human MR, do not appear to have a substantial impact on cognitive function in connection with MRI examinations.[63,64]

No correlation was observed between cognitive performance and subjectively reported effects in **Paper I**. In the present cohort, subjective effects were generally mild and of low intensity. Previous studies have reported higher levels of discomfort related to vertigo at higher field strengths; however, these effects did not appear to increase participant rejection rates, suggesting that such sensations are often transient and tolerable.[21] How participants are informed about the MRI examination and potential sensations associated with 7 T exposure may also influence how these effects are perceived and managed.[21,22] Nevertheless, potential effects in more sensitive individuals should still be taken into consideration. Further research is therefore warranted in larger and more heterogeneous populations to better understand whether cognitive performance may be secondarily affected in the presence of more pronounced perceptual responses to 7 T MRI exposure.[65]

Effects of MRI exposure on cochlear function

DPOAEs were measured to assess cochlear function before and after the MRI examination when using dual hearing protection. A total of 39 participants were included for left ear analyses and 38 for right ear analyses, as one participant had a known hearing impairment in the right ear. DPOAE responses were recorded across seven frequency bands. Values below 6 dB at baseline for a given frequency were excluded from the analyses, and missing data were primarily due to absent DPOAE responses at specific frequencies. Additional missing values occurred because some participants were unable to attend the final follow-up session, conducted several days after the MRI examinations.

A nominally significant difference was observed in the paired comparisons between baseline and the second follow-up at 2 kHz in the left ear and at 4 kHz in the right ear. However, these findings did not remain significant after Bonferroni correction. No significant differences were observed at the remaining frequencies, indicating that the overall conclusions were unaffected by the correction.

Table 9 provides an overview of mean differences, standard deviations, and effect sizes for all paired comparisons across time points and frequencies for both the left and right ears.

We did not observe any measurable impairment of cochlear function following repeated MRI examinations in **Paper II**. This is consistent with previous studies reporting limited or no detectable auditory effects after repeated MRI-related noise exposure. Although subtle changes in hearing threshold levels have been described after repeated MR acquisition, particularly within the 500–4000 Hz range in the right ear, these effects have generally been small in magnitude.[182] Similarly, another study reported isolated decreases in DPOAE amplitudes in a limited sample; however, the authors concluded that DPOAE amplitudes showed no scanner noise impact immediately after the scan session when compared with pre-scan values.[183]

These studies were conducted at 3 Tesla, and the actual SPLs to which participants were exposed remain uncertain, as measurements may not fully reflect in-bore acoustic conditions. Participants in those studies, as in our study, were protected using a combination of hearing protection devices.[182,183]

Taken together, the current findings and previous literature suggest that measurable impairment of cochlear function following MRI exposure appears limited, particularly when hearing protection strategies are applied. However, differences in field strength, hardware design and uncertainty regarding actual noise exposure should be considered when comparing results across studies.

Table 9. Paired comparisons of DPOAE measures across frequencies and timepoints.

Comparisons were performed for the left (sin.) and right (dx.) ears across measured frequencies (kHz) and timepoints. Reported *p*-values are uncorrected for multiple comparisons.

Frequency/Ear (kHz/sin. or dx.)	Baseline→ Follow-up 1		Baseline→ Follow-up 2		Baseline→Follow- up 3	
	Δmean ± SD	*t*(df), *p*	Δmean ± SD	*t*(df), *p*	Δmean ± SD	*t*(df), *p*
1 kHz/sin.	-0.59 ± 4.13	*t*(38)= -0.90, *p*=0.37	0.23 ± 5.18	*t*(37)= 0.27, *p*=0.79	0.15 ± 4.61	*t*(33)= 0.19, *p*=0.85
1.5 kHz/sin.	0.56 ± 4.06	*t*(38)= 0.86, *p*=0.39	1.43 ± 5.14	*t*(37)= 1.72, *p*=0.09	0.12 ± 3.92	*t*(33)= 0.18, *p*=0.86
2 kHz/sin.	0.86 ± 3.81	*t*(38)=1.40, *p*=0.17	1.48 ± 4.10	*t*(37)= 2.23, *p*=0.03	0.43 ± 3.88	*t*(33)= 0.65, *p*=0.52
3 kHz/sin.	0.96 ± 4.43	*t*(38)=1.35, *p*=0.19	0.73 ± 4.44	*t*(37)= 1.01, *p*=0.32	0.14 ± 4.77	*t*(33)= 0.17, *p*=0.87
4 kHz/sin.	-0.76 ± 4.59	*t*(38)= -1.03, *p*=0.31	-0.79 ± 3.42	*t*(37)= -1.43, *p*=0.16	-0.86 ± 5.10	*t*(33)= -0.98, *p*=0.33
5 kHz/sin.	-0.20 ± 4.40	*t*(38)= -0.28, *p*=0.78	0.34 ± 3.86	*t*(37)= 0.54, *p*=0.59	-0.09 ± 4.19	*t*(33)= -0.13, *p*=0.90
6 kHz/sin.	-0.53 ± 4.28	*t*(38)= -0.78, *p*=0.44	-0.45 ± 4.00	*t*(37)= -0.69, *p*=0.49	-0.35 ± 4.55	*t*(33)= -0.45, *p*=0.66
1 kHz/dx.	0.10 ± 3.94	*t*(34)= 0.15, *p*=0.88	0.33 ± 4.71	*t*(31)= 0.40, *p*=0.69	0.24 ± 5.92	*t*(30)= 0.23, *p*=0.82
1.5 kHz/dx.	-0.70 ± 4.64	*t*(37)= -0.93, *p*=0.36	0.19 ± 4.42	*t*(34)= 0.25, *p*=0.81	-0.49 ± 6.10	*t*(32)= -0.47, *p*=0.64
2 kHz/dx.	0.26 ± 4.34	*t*(37)= 0.36, *p*=0.72	0.74 ± 3.80	*t*(33)= 1.13, *p*=0.27	-0.54 ± 5.63	*t*(32)= -0.55, *p*=0.58
3 kHz/dx.	0.71 ± 5.33	*t*(36)= 0.81, *p*=0.42	0.12 ± 5.30	*t*(33)= 0.14, *p*=0.89	0.32 ± 5.11	*t*(31)= 0.35, *p*=0.73
4 kHz/dx.	-1.12 ± 5.17	*t*(37)= -1.34, *p*=0.19	-2.19 ± 5.60	*t*(34)= -2.31, *p*=0.03	-1.35 ± 5.08	*t*(32)= -1.53, *p*=0.14
5 kHz/dx.	-0.02 ± 4.51	*t*(37)= -0.03, *p*=0.98	0.73 ± 5.48	*t*(34)= 0.79, *p*=0.44	-0.52 ± 5.51	*t*(32)= -0.54, *p*=0.59
6 kHz/dx.	-0.53 ± 4.12	*t*(37)= -0.80, *p*=0.43	-1.67 ± 5.64	*t*(34)= -1.75, *p*=0.09	-0.76 ± 5.52	*t*(32)= -0.79, *p*=0.43

Building on our findings using dual hearing protection, the importance of adequate hearing protection in MRI environments should be emphasised. Although several studies have indicated that MRI-related noise exposure is not associated with measurable auditory effects,[110,111,184,185] others have reported subtle auditory changes. These include frequency-specific shifts in hearing thresholds, even when passive hearing protection is used.[106,114,164,165,186] In particular, noise-induced hearing loss in an unobstructed ear typically affects frequencies around 4 kHz, where the human auditory system is most vulnerable to acoustic damage.[187,188]

Against this background, appropriate and well-functioning hearing protection is of critical importance during MRI examinations. It is therefore essential that hearing protection is carefully selected, correctly fitted, and functionally verified before the start of the examination. Given that protection efficacy is highly dependent on both seal and placement, particular attention should be paid to ensuring an optimal fit for each participant.[189] The use of dual hearing protection is also recommended, as evidence indicates significantly improved attenuation compared with single protection.[190]

Furthermore, it is equally important to ensure that hearing protection maintains its effectiveness throughout the entire MRI procedure. Movement, discomfort, or improper positioning during scanning may compromise attenuation levels, potentially increasing auditory exposure. Continuous attention to participant comfort and protection integrity is therefore warranted to ensure consistent hearing protection and minimise the risk of noise-induced auditory effects.

Dental putty has occasionally been used for hearing protection during MRI; however, its use is generally discouraged due to uncertain protective performance and the potential risk of material fragments detaching and entering the ear canal.[69] In the present study, an alternative impression material intended for hearing aid moulds was used, but only in combination with a foam earplug.[82,154,163] Its primary function was to stabilise the earplug, with a secondary role in providing additional attenuation, and it should never be used as the sole means of hearing protection.

Future studies should examine the effects of different scanning protocols, sound pressure levels, and individual susceptibility to noise-induced auditory outcomes. Extending auditory safety studies to newer systems with stronger gradients, such as the NextGen at Berkeley,[191] the 7 T MAGNETOM Terra.X,[192] the GE Signa 7 T,[193] and the 3 T MAGNETOM Cima.X,[194] broadens the relevance of previous findings and ensures that they remain applicable to current MRI technology.

Evaluation of software-based ANR at 7 Tesla

This section presents the evaluation of software-based ANR at 7 T. The analysis included measurements of SPLs and PNS, participants' self-reported perceptions of noise and PNS, and subjective and objective assessments of image quality. Imaging performed with and without the SofTone technique was compared.

Sound pressure and PNS levels

SPLs measured across all participants for each sequence showed significantly lower acoustic levels for the SofTone sequences compared with the conventional sequences. This effect was observed for both T_2W FSE and 3D T_1W TFE sequences (Table 10). For T_2W FSE, the mean difference in peak SPLs was 19.3 dBA, corresponding to an 89% reduction. For 3D T_1W TFE, the corresponding values were 22.2 dBA and a 92% reduction.

Table 10. Peak SPLs for conventional and SofTone sequences

Sequence	Conventional Mean ± SD, dBA	SofTone Mean ± SD, dBA	*p*-value
T_2W FSE	116.3 ± 1.8	97.0 ± 2.1	**<0.001**
3D T_1W TFE	123.7 ± 1.2	101.5 ± 1.2	**<0.001**

Additionally, a significant reduction in scanner-derived PNS levels was observed for SofTone sequences compared with conventional acquisitions across T_2W FSE and 3D T_1W TFE sequences (Table 11). For T_2W FSE, the median difference in PNS levels was 47.5 percentage points. For 3D T_1W TFE, the corresponding median difference was 67 percentage points.

Table 11. PNS levels for conventional and SofTone sequences

Sequence	Conventional Median (IQR; range)	SofTone Median (IQR; range)	*p*-value
T_2W FSE	67 (17; 50–70)	20 (1; 18–24)	**<0.001**
3D T_1W TFE	83 (3; 79–84)	16 (0; 0)	**<0.001**

Self-reported noise and PNS perception

SofTone sequences were perceived as significantly quieter than the corresponding conventional acquisitions. On the Borg CR10 scale®, conventional T_2W FSE was rated close to "strong/loud" (4.5 ± 1.7), whereas T_2W FSE SofTone was rated closer to "moderate" (3.1 ± 1.1). Conventional 3D T_1W TFE was rated between "strong"

and “very strong” (5.7 ± 2.0), while the SofTone sequence was rated between “moderate” and “strong” (3.9 ± 1.7).

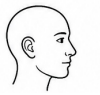
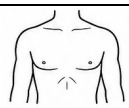
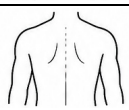
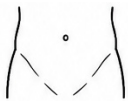
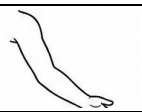
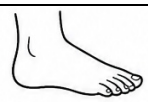
Regarding self-reported experiences of PNS, no statistically significant differences were observed between the conventional and SofTone variants in terms of perceived presence, intensity, or unpleasantness (Table 12). However, PNS was reported more frequently during the conventional sequences than during the corresponding SofTone acquisitions. Sixteen of the 28 participants reported PNS during the conventional 3D T_1W TFE sequence, compared with ten during the SofTone variant. For the T_2W sequence, the corresponding numbers were 20 and 17 participants, respectively. Ratings of unpleasantness for the conventional 3D T_1W TFE sequence approached statistical significance.

Table 12. Self-reported PNS for conventional and SofTone sequences.

Results are reported as median ratings, with interquartile range (IQR) and minimum–maximum range presented in parentheses.

Outcome	T_2W FSE			3D T_1W TFE		
	Conventional	SofTone	*p*	Conventional	SofTone	*p*
	Median (IQR; range)			Median (IQR; range)		
Presence of PNS	2 (2; 1-4)	2 (1; 1-4)	0.31	2 (2; 1-6)	1 (2; 1-5)	0.07
Intensity of PNS	2 (1; 1-4)	2 (1; 1-4)	0.52	2 (2; 1-4)	1 (2; 1-5)	0.13
Discomfort of PNS	2 (1; 1-5)	2 (1;1-3)	0.15	2 (2; 1-6)	1 (1; 1-5)	0.05

For the T_2W FSE sequence, most participants (14 participants) reported PNS sensations in the upper body, hand, and arm, while eight participants reported the same regions for the SofTone variant. A similar distribution was observed for the conventional 3D T_1W TFE sequence, where eleven participants reported PNS in the upper body, hand, and arm. In contrast, only two participants reported these regions for the SofTone variant. Interestingly, a substantial proportion of participants (7 participants) reported sensations in their legs during the 3D T_1W SofTone sequence (Figure 7).

BODY REGIONS		T_2W FSE		3D T_1W TFE	
		Conventional	SofTone	Conventional	SofTone
	Head	0	3(11%)	0	0
	Eye	0	1(4%)	0	0
	Upper body	2(7%)	2(7%)	6(21%)	1(4%)
	Back	0	1(4%)	1(4%)	0
	Abdomen	0	1 (4%)	0	0
	Pelvis	0	0	0	1 (4%)
	Arm	6(21%)	3(11%)	4(14%)	0
	Hand	6(21%)	3(11%)	1(4%)	1(4%)
	Leg	4(14%)	3(11%)	3(11%)	7(25%)
	Foot	2(7%)	0	1(4%)	0

Number of participants who reported PNS (*n*)

0	1-2	3-5	6-10

Figure 7. Distribution of PNS across body regions
The figure shows the distribution of PNS across body regions during conventional and SofTone MRI sequences. Values represent the number of participants reporting PNS in each body region, with percentages of the total sample (*n*=28) presented in parentheses.

The present findings regarding perceived noise levels are consistent with previous studies demonstrating that participants perceive ANR sequences as quieter than conventional MRI sequences.[32,79,119,120,132] In addition to reduced noise levels, earlier research has also reported that participants described the acoustic characteristics of ANR sequences as more pleasant to listen to.[132]

Previous research has also demonstrated that allowing participants to listen to music during MRI examinations can improve patient comfort, reduce anxiety, and increase overall satisfaction with the examination experience.[30,79,195] However, the high acoustic noise levels generated during MRI often interfere with the audibility and perceived quality of the music, thereby limiting its beneficial effects. In this context, acoustic noise reduction techniques may provide additional advantages beyond simple noise attenuation.

Together, these findings suggest that acoustic noise reduction techniques such as ANR and SofTone may contribute not only to lower perceived loudness but also to an overall more comfortable examination experience.

Additionally, PNS was more common during the conventional sequences compared to the SofTone variant, although most reports reflected low-intensity sensations. Participants were informed about possible PNS sensations and reassured regarding their safety before scanning, which may have influenced subjective ratings. Nevertheless, this highlights the importance of clear participant information for reducing anxiety and improving tolerance during MRI procedures,[30,127] while further supporting the potential of ANR techniques to enhance comfort during UHF MRI examinations. Furthermore, previous research has demonstrated that anatomical and physiological differences, including body composition and tissue distribution, may influence individual PNS thresholds and subjective perception.[92] This variability may partly explain why higher scanner-derived PNS levels did not correspond to significantly higher participant-reported discomfort in the present study.

The distribution of reported PNS was somewhat unexpected. Previous studies conducted on the same MRI system (7 T facility in Lund) have shown that PNS most commonly affects the torso, hand, and arm during head-first examinations, consistent with expected current pathways induced by gradient switching.[23] However, in a previous study, it has also been reported that more than 15% of participants experienced PNS in the lower extremities in head-first acquisitions, although upper limb involvement remains more frequent, affecting just over 20% of cases.[20] In **Paper IV**, participants had no prior personal experience of how PNS is perceived during MRI, although a verbal description was provided before scanning. This does not rule out that some reported sensations were unrelated to PNS, but instead caused by scanner vibrations, transient extremity paraesthesia, or similar sensations.[77,196-198] Such factors may have influenced the reported distribution patterns and highlight the inherent challenge of distinguishing true PNS from other sensory phenomena in subjective reporting.

Image quality

SNR was compared between the conventional and SofTone sequences in both grey and white matter. In grey matter, SNR was significantly higher in the conventional sequence compared to the SofTone sequence, corresponding to a relative increase of 15.9%. SNR in white matter was higher in the conventional sequence, corresponding to a relative increase of 19.3%, but the difference was not statistically significant. However, considerable inter-subject variability was observed, as reflected by the wide range of relative differences. Furthermore, a higher CNR was also observed for the conventional sequence compared to the SofTone sequence, with a relative increase of 34% (Table 13).

Table 13. Comparison of SNR and CNR between conventional and SofTone MRI sequences (*n*=28).

Measure	Sequence		*p*-value
	Conventional 3D T_1W TFE	SofTone 3D T_1W TFE	
SNR grey matter	18.5 ± 5.05	16.3 ± 4.27	**0.024**
SNR white matter	29.7 ± 7.20	26.4 ± 6.95	0.068
CNR	7.10 ± 1.91	5.86 ± 2.17	1.00

Subjective image evaluation further indicated reduced grey–white matter differentiation in the 3D T_1W TFE sequence, as two of the three readers graded the sequence lower compared to the conventional acquisition. Table 14 presents the distribution of the image quality ratings (mean ± SD) for the 3D T_1W TFE sequences assessed by the three reviewers. The grading scale ranged from 1 to 4, where 1 corresponded to poor image quality and 4 to excellent image quality.

In contrast, the T_2W FSE sequences showed slightly less interobserver variability among the neuroradiologists in the assessment of the conventional and SofTone sequences. Furthermore, no significant differences were observed in overall image quality, grey–white matter differentiation, or CSF–soft tissue differentiation (*p*-values ranging from 0.21 to 1.00).

Table 14. Image quality scores for each neuroradiologist across MRI sequences (*n*=26)

The distribution of ratings is expressed as mean ± SD for each MRI sequence as assessed by the three neuroradiologists. *p*-values are reported for comparisons between conventional and SofTone sequences within each reader.

		Sequence		
Readers	Image quality criteria	Conventional 3D T_1W TFE	SofTone 3D T_1W TFE	*p*-value
Reader 1	General image quality	3.65 ± 0.56	3.08 ± 0.63	**<0.01**
	Grey-white matter differentiation	3.85 ± 0.37	3.62 ± 0.57	0.06
	CSF–soft tissue differentiation	4.00 ± 0.00	3.96 ± 0.20	0.32
Reader 2	General image quality	3.04 ± 0.34	2.85 ± 0.54	0.13
	Grey-white matter differentiation	3.54 ± 0.58	3.19 ± 0.63	**0.02**
	CSF–soft tissue differentiation	3.77 ± 0.43	3.50 ± 0.51	0.05
Reader 3	General image quality	2.73 ± 0.45	2.50 ± 0.51	0,08
	Grey-white matter differentiation	2.81 ± 0.40	2.27 ± 0.45	**<0.001**
	CSF–soft tissue differentiation	3.00 ± 0.00	2.88 ± 0.33	0.08

Reader 2 rated the T_2W FSE SofTone sequence as having significantly fewer artefacts than the conventional sequence. For the remaining assessments, no statistically significant differences in artefact levels were observed between sequences acquired with and without SofTone (Table 15).

The qualitative image assessment revealed a consensus among all three readers that image quality could be rated as good or very good within individual sequences; however, a localised reduction in image quality in specific regions often resulted in a lower overall score. It was further noted that the observed artefacts were predominantly 7 T-specific in nature. In addition, there was some variability in how these artefacts were described and interpreted between readers, suggesting a degree of subjectivity in artefact classification and attribution.

Table 15. Artefact ratings across MRI sequences (*n*=26)

Data are presented as mean ± SD for each sequence and reader, with p-values for within-reader comparisons between conventional and SofTone.

Readers	Conventional T_2W FSE	SofTone T_2WFSE	*p*	Conventional 3D T_1W TFE	SofTone 3D T_1W TFE	*p*
Reader 1	3.12 ± 0.43	3.12 ± 0.52	1.00	2.92 ± 0.39	2.77 ± 0.43	0.16
Reader 2	2.88 ± 0.43	3.04 ± 0.45	**<0.05**	3.00 ± 0.28	2.88 ± 0.43	0.18
Reader 3	2.77 ± 0.43	2.73 ± 0.45	0.74	2.88 ± 0.33	2.69 ± 0.55	0.13

Notably, even in cases rated as exhibiting major artefacts, CSF differentiation was often preserved at an excellent level. This may be explained by the spatial distribution of artefacts, for example, when motion-related artefacts were located peripherally or confined to one side of the image, thereby sparing central anatomical structures such as the ventricular system.

Susceptibility artefacts were primarily observed in regions adjacent to the paranasal sinuses and skull base. Although the FOV did not always include the nasal sinuses in all T_2-weighted acquisitions, and therefore these artefacts were not consistently assessed in those cases, susceptibility effects remain a well-known and persistent challenge at 7 T in these anatomical regions.[28]

Although ANR techniques have consistently demonstrated substantial reductions in MRI noise levels,[32,119-121,132,199] direct comparisons between studies remain challenging. This is largely due to differences in study design, MRI systems, acquisition parameters, and the specific ANR implementations used. Furthermore, the adjustments made between conventional and ANR-modified sequences vary considerably across studies, making it difficult to establish standardised comparisons regarding both acoustic performance and image quality outcomes.

Nevertheless, several studies have reported that image quality obtained with quieter MRI sequences can be maintained at an acceptable to high level.[32,119,131,132,199] PETRA (Pointwise Encoding Time Reduction with Radial Acquisition) is an MRI technique that reduces acoustic noise through acquisition strategies involving minimal gradient switching. In some cases, the technique has also shown advantages in aspects of image quality compared with conventional imaging sequences.[200,201] Furthermore, recent technical MR research investigates novel gradient design approaches, including silent ultrasonic gradient systems, with the potential to reduce acoustic noise and PNS while preserving imaging performance.[202,203]

Despite these promising findings, it remains essential to evaluate potential trade-offs between acoustic noise reduction and image quality in relation to the intended purpose of the examination. ANR techniques should therefore be implemented with consideration[120,121] of their possible influence on image characteristics, diagnostic performance, and quantitative image analysis.

The present thesis should be viewed as an initial step toward establishing knowledge regarding the ANR methods available on the current MRI system. In addition to evaluating their suitability for morphological assessment in clinically oriented examinations, the study also aims to investigate how image properties may affect subsequent volumetric analyses used in fMRI-related research.

Radiographer's role and interprofessional collaboration

Radiographers play a pivotal role in implementing safe and effective MRI protocols, combining technical expertise with a person-centred approach. Within multidisciplinary teams, this facilitates the systematic evaluation of physiological and perceptual effects while minimising adverse outcomes without compromising image quality, highlighting the importance of interprofessional collaboration. In parallel, the integration of knowledge from fields such as medicine, physics, and neuroscience reflects the interdisciplinary nature of 7 T MRI research.

Person-centred, compassionate care must be a foundation of practice within the radiography profession alongside technological advancements.[204] A recent study on person-centred care in radiography describes a "skewed balancing act" between technical demands and patient interaction, where technical priorities often dominate.[140] This is particularly evident in the role of the MRI research radiographer, who works to ensure that advanced imaging protocols are delivered safely, effectively, and with sustained attention to participant well-being.[139] Recent literature highlights their key contribution to participant safety, ethical conduct, and the implementation of advanced imaging. It also identifies challenges related to resources, interdisciplinary communication, and ethical complexities, such as incidental findings.[205]

Although interdisciplinary communication can be challenging, it is essential, as it enables multiple perspectives to be applied to a given problem, thereby facilitating the development of more well-defined, feasible, and valuable research questions.[206] Within such settings, radiographers contribute a unique and person-centred perspective, grounded in their proximity to the clinical encounter, bringing experiential knowledge not always accessible to more distant professional groups.[137] At the same time, other disciplines provide equally indispensable expertise and specialised knowledge, and it is the integration of these complementary perspectives that strengthens both research quality and clinical relevance.

Building on a quote attributed to Benjamin Franklin,*"tell me and I forget. Teach me, and I remember. Involve me, and I learn",* this work emphasises the importance of multidisciplinary collaboration. Through active involvement across disciplines, mutual learning is fostered, allowing research to be developed, refined, and evaluated from multiple perspectives, ultimately benefiting patient care.[207]

Methodological considerations, strengths and limitations

A methodological strength of the studies included in this thesis was the interdisciplinary collaboration between radiographers, radiologists, audiologists, psychologists, physicists, and engineers. Integrating expertise from multiple disciplines facilitated a broader understanding of both human responses and technical performance in 7 T MRI. It also reduced the risk of a narrow, single-discipline focus, thereby enabling a more comprehensive evaluation of safety, comfort, and imaging quality.

In addition to strengthening the methodological framework of the studies, the interdisciplinary collaboration also constituted an important learning process throughout my doctoral education. Working within this multidisciplinary environment provided opportunities to develop my competence in new measurement and analysis methods, including OAE measurements and quantitative image analysis. It also facilitated a deeper understanding of the radiological perspective through participation in qualitative image assessment. Furthermore, it enabled hands-on experience in the evaluation and improvement of passive hearing protection, the development and optimisation of MRI protocols for noise reduction, and the planning and collection of research data. Collectively, these experiences contributed not only to the research presented in this thesis but also to the broader educational aim of doctoral training by fostering methodological breadth and interdisciplinary experience for future research.

A limitation of the present work is the exclusive reliance on quantitative study methods. While this approach enables objective measurement and statistical comparison of outcomes, it may not fully capture the complexity of participants' subjective experiences during MRI examinations. Qualitative research could have provided a more nuanced understanding of how individuals perceive and interpret factors[208] such as noise exposure, comfort, and anxiety in the scanner environment. Previous qualitative interview findings have suggested that acoustic noise, together with other patient-related and contextual factors, may contribute to a "tipping point" leading to premature termination of MRI examinations.[29] This further supports the value of qualitative approaches for exploring patient experiences.

Furthermore, based on clinical experience with MRI examinations over several years, it has become apparent that patients may report experiences that are not easily captured through measures of statistical significance alone. Such experiences may include subtle discomfort, situational anxiety, or sensory overload, which can vary considerably between individuals[65] and may not be fully reflected in quantitative outcome measures.

The studies were conducted in healthy participants with a relatively young sample in **Papers I** and **II** (age range 18–40 years) and 18–56 years in **Papers III** and **IV**. For the cognitive assessments, this was a strength, as it may have minimised the

influence of age-related cognitive decline on measures such as processing speed and visuospatial function.[209,210] However, the limited age range may reduce the generalisability of auditory findings to more vulnerable populations with increased sensitivity to acoustic exposure, such as older adults and clinical patients. Individuals with cochlear impairment or increased auditory vulnerability may experience elevated loudness already at or near the hearing threshold (i.e., the lowest sound level that is detectable), which can influence how scanner noise is perceived even at comparable sound levels.[211] This underscores the importance of considering not only objective sound intensity, but also individual differences in loudness perception when evaluating participant experience in MRI environments.

Furthermore, participant recruitment was based on voluntary participation, with individuals self-enrolling in the study without any clinical indication. This voluntary nature may have influenced participants' initial mindset and attitudes toward the examination, potentially resulting in a more relaxed and less anxious disposition compared to a clinical patient population. Consequently, the findings may reflect experiences shaped by a self-selected group who may differ from patients undergoing MRI for clinical reasons.

A general methodological consideration in the selection of assessment tools in this thesis was to use methods that could be administered in close temporal proximity to the MRI examination. This was done to minimise the risk that any effects of the MRI environment would have dissipated, and to ensure that potential short-term influences of the scanning situation were captured without excessive delay. In addition, the assessment procedures were designed to impose as little time burden as possible on participants, without compromising the quality or relevance of the data obtained. Furthermore, the selected methods needed to be easily implementable in terms of handling and administration, while also being well established within the research field.

A limitation in **Paper I** is that the cognitive testing was only conducted before and after the MRI examination, and not during the scanning itself. As a result, potential transient effects with rapid recovery may have been overlooked. In this context, the paper-and-pencil DSST offers several practical benefits compared to computerised tools. The DSST requires only a short administration time (approximately 2 minutes), making it considerably quicker than many computerised tests. Although the task could have been administered digitally, a paper-and-pencil format was chosen to incorporate an element of basic manual dexterity.[151]

DSST is a clinically useful cognitive test due to its sensitivity to cognitive deficits across a wide range of brain diseases and conditions.[147,148,212,213] However, without further testing, the DSST does not provide information about the specific nature of the deficit.[151] From a methodological perspective, reliance on a single, global cognitive measure may limit the ability to detect more subtle or domain-specific effects potentially associated with 7 T MRI exposure.

The inclusion of a more comprehensive cognitive test battery could have strengthened the assessment by enabling a more detailed characterisation of potential effects across different domains. Such an approach might include tests targeting eye–hand coordination (e.g., Pursuit Aiming II or adapted line-bisection tasks), working memory (e.g., WAIS-III or N-back), and visual processing (e.g., VISTECH contrast sensitivity or tracking tasks).[124] However, the addition of a broader test battery must be balanced against practical considerations, such as increased testing time and potential fatigue effects, particularly in the context of MRI examinations.

Two different methods were employed for the measurement of sound levels in this thesis. In **Paper II**, a more comprehensive sound measurement approach was used, capturing a broader range of SPLs. This was important to obtain an accurate overview of the acoustic exposure experienced by participants, as the primary aim of the study was to investigate potential effects on hearing. In contrast, **Paper III** focused solely on measuring peak SPLs. While a more extensive acoustic assessment could also have been valuable in this context, the objective was to compare noise levels between different MRI sequences rather than to evaluate the effects of overall sound exposure. Nevertheless, measuring peak levels is relevant, as a single high-intensity noise exposure exceeding 120 dB(A) may result in permanent hearing damage.[45]

In relation to the assessment of auditory effects associated with 7 T MRI exposure, OAE measurements were selected as the primary outcome measure. This choice was guided by their frequent use in previous studies investigating MRI-related auditory effects, as well as their established role in evaluating cochlear function.[106,110,111,164,165,185,214] DPOAEs primarily reflect the functional integrity of outer hair cells and constitute an objective, non-invasive, cost-effective, and time-efficient method for assessing cochlear status.[166]

Furthermore, DPOAEs have been shown to correlate with pure-tone audiometry and speech-evoked auditory brainstem response measures[215,216] and may detect early subclinical changes in cochlear function before their manifestation in pure-tone audiometry.[217] In addition, since the DPOAE test is an objective measure that does not require any behavioural response from participants, fatigue is unlikely to have a significant impact on the outcome.[183] Considering these characteristics, the DPOAE test was therefore considered the preferred method for evaluating hearing status in **Paper II**.

Previous studies have demonstrated that DPOAEs show particularly good performance in the 2–8 kHz frequency range, which corresponds to frequencies commonly affected by noise exposure.[218] In our study, DPOAE measurements were obtained across 12 frequencies ranging from 0.5 to 10 kHz; however, only the 1–6 kHz frequency range was included in the final analyses. This decision was based on known methodological constraints associated with OAE recordings at the spectral

extremes. Low-frequency emissions are characterised by reduced reproducibility due to increased susceptibility to background noise contamination and variations in middle ear pressure, which may compromise measurement reliability.[165] Conversely, higher-frequency responses typically demonstrate elevated thresholds and increased inter-subject variability, thereby reducing signal stability.[219] Consequently, restriction to the 1–6 kHz range was considered to optimise measurement robustness and ensure comparability with previous literature.[165]

The present work (**Papers III** and **IV**) employed a software-based optimisation approach aimed at smoothing steep gradient current transitions, with the intention of reducing acoustic noise generation in UHF MRI. This choice is grounded in the fact that MRI acoustic noise depends not only on magnetic field strength but also on gradient strength, which is elevated in higher-field systems to reduce artefacts. Consequently, both factors contribute to higher SPLs, with doubling of either parameter increasing SPLs by approximately 6 dB(L), and simultaneous increases resulting in an additive effect.[220] These mechanisms highlight that acoustic noise at 7 T is closely linked to gradient-related sequence design, and that optimisation of gradient implementation therefore represents a relevant methodological strategy for noise reduction.

Conversely, a methodological consideration is that the SofTone-optimised sequences require increases in both TE and receiver bandwidth, parameters known to influence SNR and CNR. Increasing TE generally reduces SNR due to increased transverse magnetisation decay, while increasing receiver bandwidth increases image noise and thereby lowers SNR. These parameters may also indirectly affect CNR through alterations in tissue contrast and noise characteristics.[27] Consequently, some degree of change in quantitative image quality was expected. However, it remains important to evaluate the magnitude and practical implications of these effects, particularly given that variations in image characteristics may influence structural image-processing pipelines commonly employed in neuroimaging analyses.[221]

Alternative approaches, such as zero echo time (ZTE) imaging, enable very efficient and nearly silent MRI scanning. Signal encoding starts immediately after excitation, resulting in an effective echo time of zero and allowing noticeably short repetition times. Without the need to switch off the gradients between TR intervals, the gradients can remain continuously active, thereby greatly reducing acoustic noise.[222,223] ZTE Silent imaging has shown promising results for high-contrast T_1-weighted imaging at 7 T, including reduced acoustic noise, improved homogeneity, and absence of flow artefacts.[136] However, this technique was beyond the scope of the present thesis and should be explored in future research.

In this thesis, the evaluation of ANR sequences was based on both qualitative and quantitative image assessment. This dual approach was considered complementary, with qualitative evaluation capturing overall image interpretability, while

quantitative analysis enables an objective assessment of image quality metrics. As a first step in the systematic evaluation of ANR imaging at the 7 T facility, the work also provides a preliminary basis for exploring whether ANR sequences may offer a more tolerable or participant-friendly imaging approach.

In the qualitative image assessment, the overall image was evaluated based on the presence of artefacts, general image quality, the ability to differentiate between white and grey matter, and the distinction between CSF and surrounding soft tissue. No specific anatomical regions were predefined for the evaluation, meaning that the assessment was based on a global impression of the image quality. A general observation among all three readers was that images could be rated as good or very good overall within a given sequence, while the visibility of one or more specific anatomical areas was sometimes suboptimal, which in turn contributed to a lower overall score.

The quantitative evaluation of image quality in ANR sequences was considered particularly relevant given the central role of structural MRI acquisitions in neuroimaging research. Structural sequences such as 3D T_1W imaging are routinely used to visualise neuroanatomy and to enable anatomical localisation in both clinical and research settings.[221] Consequently, ANR-related alterations in image characteristics may have implications extending beyond visual image interpretation alone.

Furthermore, one reader noted that certain images, although visually acceptable, might still pose challenges for subsequent analysis using volumetric software. This highlights a potential discrepancy between visual assessment and requirements for quantitative post-processing. To address this, quantitative image analysis was also included to evaluate the extent to which the implementation of ANR sequences may influence image properties relevant for subsequent structural image-processing procedures. This was motivated by the anticipated impact of increased TE and receiver bandwidth in the SofTone sequence on key image quality metrics.

The evaluation of quantitative image quality was primarily limited to the most relevant quantitative parameters for the study purpose, namely SNR and CNR. Other image quality metrics, such as spatial resolution, geometric distortion, and artefact assessment, were considered less central. This was because the imaging protocols were otherwise kept constant, and the primary focus was on relative differences between the two approaches rather than a comprehensive technical optimisation.

Conclusions

This thesis demonstrates that UHF (7 T) MRI involves a complex interplay between physiological effects, participant experience and technical performance. No consistent detrimental effects on cognitive performance were observed following 7 T MRI exposure beyond transient learning-related variations. At the same time, subjective sensations such as dizziness underline the importance of participant monitoring during and after scanning.

Auditory safety assessments showed no measurable changes in outer hair cell function following cumulative exposure to 7 T MRI noise when dual hearing protection was used. This supports its effectiveness as a protective strategy in both clinical and research settings.

Technical optimisation strategies were shown to substantially improve the scanning environment. The use of gradient software-optimised ANR (SofTone) significantly reduced SPLs and was perceived as more comfortable by participants, while maintaining acceptable subjective image quality. Similarly, ANR strategies reduced PNS, improving objective safety margins, although this was accompanied by modest but measurable reductions in quantitative image quality metrics (SNR and CNR), highlighting an inherent trade-off in sequence optimisation.

Overall, the findings emphasise that safe and effective implementation of 7 T MRI requires a balanced approach that integrates technical performance with physiological safety and participant comfort. While advanced sequence and noise-reduction strategies can substantially improve tolerability, careful consideration of their impact on image quality remains essential. Collectively, this work contributes to a more comprehensive understanding of how UHF MRI can be optimised to support both diagnostic performance and human well-being.

Clinical implications

When placing these results in a broader context, it is important to consider current safety guidelines and regulatory limits for MRI, including those related to acoustic noise exposure, gradient switching, and PNS thresholds. The observed effects on perceived noise and auditory function underline the relevance of maintaining adequate hearing protection and monitoring SPLs during scanning. Similarly, the assessment of PNS in relation to sequence design reflects the importance of adhering to established safety limits while optimising imaging protocols. In this context, the findings support the need for continued evaluation of how UHF MRI operates within existing regulatory frameworks.

The results further illustrate the inherent trade-offs involved in MRI optimisation. While advanced sequences and high gradient performance can improve image quality, they may also increase acoustic noise and the likelihood of PNS. Conversely, technical strategies such as SofTone demonstrated potential to reduce perceived noise and PNS but require careful evaluation to ensure that image quality remains sufficient for diagnostic and research purposes. These trade-offs highlight the necessity of balancing competing demands, where improvements in one domain may influence another.

From a clinical and practical perspective, these findings underscore the importance of integrating person-centred considerations into MRI protocol development and adapting imaging protocols according to the purpose of the examination. The combination of subjective assessments, such as perceived noise, and objective measurements, including DPOAEs and PNS evaluation, provides a more complete basis for decision-making. This approach allows for the identification of strategies that not only comply with safety standards but also enhance participant comfort and tolerance during scanning.

Within this framework, the radiographer plays a key role in bridging the technical and human aspects of MRI. By applying knowledge of MRI physics, safety regulations, and individual variability in responses, radiographers contribute to the optimisation of imaging protocols in both research and clinical settings. The results of this thesis demonstrate how such an integrated approach can support the ongoing development of 7 T MRI, ensuring that advancements in image quality are aligned with considerations of safety and individuals' well-being.

Future perspectives

Building on the findings and methodological insights gained from the present studies, several directions for future research can be identified. Future research should further explore the interaction between UHF MRI and human physiological and perceptual responses under more ecologically valid conditions. Studies that assess cognitive and sensory effects during MRI acquisition, rather than relying solely on pre- and post-scan measurements, may provide a more accurate understanding of transient and real-time responses to the scanning environment.[63]

Additionally, qualitative interviews could be used to deepen the understanding of subjective human experiences during UHF MRI examinations, including discomfort, anxiety, and coping strategies. This would complement quantitative assessments by capturing aspects of the experience that are not accessible through structured measures. It would also help formalise clinically observed, but often undocumented, experiential knowledge, often gained by radiographers during close interaction with participants.

Further work is also needed to optimise acoustic noise reduction strategies and gradient waveform modifications to minimise trade-offs between participant safety, comfort, and image quality. It would also be of interest to evaluate noise-cancelling devices as complementary approaches for reducing acoustic noise exposure.

While current approaches such as SofTone demonstrate clear benefits in reducing SPLs and PNS, their impact on quantitative imaging metrics suggests that sequence optimisation remains an important area for continued development. Furthermore, longitudinal and repeated-measures studies may provide deeper insights into adaptation effects and potential cumulative impacts of 7 T MRI exposure.[224]

In addition, future studies should expand auditory safety assessments to include more diverse populations and longer-term exposure paradigms, as well as individuals with pre-existing hearing vulnerability. This would improve the generalisability of current findings regarding hearing protection effectiveness.

While the present work did not investigate the effects of ANR sequences on downstream post-processing procedures used in neuroimaging research, this represents an area of potential interest for future studies. Further investigations may therefore benefit from evaluating whether ANR-related alterations in image characteristics influence structural image-processing pipelines commonly employed

in neuroimaging analyses.[221] This may be particularly relevant for widely used morphometric approaches such as voxel-based morphometry, where subtle variations in image contrast and tissue classification may affect the detection of regional grey matter differences between groups or in relation to covariates. In addition, future studies could investigate the potential impact of ANR sequences on automated tissue-type segmentation of grey matter, white matter, and cerebrospinal fluid, as well as on the segmentation accuracy of deep grey matter structures such as the hippocampus and thalamus.

There is also a need for further methodological development and evaluation of other sequence types, particularly EPI sequences characterised by rapid gradient switching and substantial acoustic noise generation.[78] This is especially relevant given the widespread use of EPI in fMRI and neuroimaging research, where prolonged acquisition times may increase participant discomfort and acoustic exposure, as well as introduce non-task-related neural activity during task performance.[38,39]

The transition from doctoral researcher to independent investigator enables a shift toward initiating and leading research projects. Future work will therefore not only build upon existing datasets and collaborations but also involve the development of new research questions related to MR safety, human perception in UHF environments, and methodological innovation in UHF MRI.

A key continuation of this research involves completing and further developing an ongoing collaboration with the 7 T facility in Trondheim, Norway, which was initiated early during the doctoral studies but delayed due to the COVID-19 pandemic. The primary aim of this collaboration is to compare participants' subjectively perceived effects during 7 T MRI across two independent sites, using harmonised questionnaire-based assessments. By examining both consistencies and differences between cohorts, the project seeks to identify factors that may influence individual experiences, such as sensations of dizziness during passage through the bore. This comparative approach enables investigation of site-specific as well as individual determinants of perceptual responses in UHF MRI environments.

From a broader perspective, continued engagement in international and Nordic research networks will be essential for advancing the field. Active participation in collaborative groups focused on 7 T MRI safety can facilitate multicentre studies, standardisation efforts, and the development of consensus-based recommendations.

Furthermore, objective physiological measures could complement subjective reports of participants' experiences during 7 T MRI examinations. For example, electromyography (EMG) and electrodermal activity (EDA) could be used to assess muscular tension and autonomic nervous system responses, providing insights into individual stress and arousal levels. EMG allows for monitoring of subtle muscle activity associated with discomfort or startle responses, while EDA reflects changes in sweat gland activity linked to arousal. These measures could help quantify

individual reactions to scanner noise, evaluate the effectiveness of hearing protection, and further improve our understanding of the physiological impact of UHF MRI environments.

As many biological and physiological effects in MRI are influenced not only by static magnetic field strength but also largely by time-varying magnetic fields, further studies investigating systems with substantially higher gradient performance are warranted.[194] Such investigations may provide additional insight into the relationship between gradient-induced effects, participant experience, and image quality beyond what can be inferred from field strength alone.

Populärvetenskaplig sammanfattning

Magnetresonanstomografi (MR) är en medicinsk bilddiagnostisk metod som används för att skapa detaljerade bilder av kroppens inre utan att använda joniserande strålning. En vidareutveckling av tekniken är ultrahöga magnetfältstyrkor som 7 Tesla MR, som ger betydligt högre bildkvalitet och bättre detaljupplösning jämfört med kliniska MR-system med lägre fältstyrka. Detta gör tekniken särskilt värdefull inom hjärnforskning, men den har även en snabbt växande klinisk användning.

Samtidigt innebär det starkare magnetfältet och den mer avancerade tekniken vid 7 Tesla MR också vissa utmaningar. Människor som undersöks i MR-kameran kan uppleva tillfälliga effekter såsom yrsel, obehag eller sensoriska förändringar. Dessutom genererar de snabbt växlande gradientfälten mycket höga ljudnivåer, vilket kan påverka upplevelsen av undersökningen och väcka frågor kring hörselskydd, särskilt vid upprepade exponeringar. En annan aspekt vid högfälts-MR är så kallad perifer nervstimulering. Det kan kännas som små muskelryckningar eller pirrningar under själva undersökningen och beror på hur magnetfältet i MR-kameran påverkar kroppens nerver under korta ögonblick. Även om känslan är ofarlig och övergående kan den bidra till obehag under undersökningen.

Trots att dessa effekter är kända finns det fortfarande kunskapsluckor kring hur människor påverkas av 7 Tesla MR-miljön och hur dessa effekter kan minskas utan att försämra bildkvaliteten. Det är även oklart i vilken utsträckning upprepad ljudexponering, trots hörselskydd, kan påverka innerörats funktion över tid. Vidare behövs mer kunskap om hur tekniska lösningar för bullerreduktion och minskad perifer nervstimulering påverkar både upplevelse och bildkvalitet.

Det övergripande syftet med denna avhandling var därför att undersöka fysiologiska och upplevelsemässiga effekter i samband med 7 Tesla MR samt studera metoder som kan göra undersökningen mer behaglig och skonsam utan att försämra bildkvaliteten.

Avhandlingen omfattar fyra delstudier. Den första studien undersökte kognitiv prestation och subjektiva upplevelser i samband med 7 Tesla MR-undersökning. Den andra studien analyserade om upprepad exponering för 7 Tesla MR påverkar innerörats funktion hos friska vuxna vid användning av dubbla hörselskydd. Den tredje studien utvärderade en ljudreducerande teknik (SofTone) och dess påverkan på ljudnivå, upplevt buller samt hur bildkvalitén bedömdes av neuroradiologer. Den

fjärde studien undersökte hur SofTone påverkar perifer nervstimulering, upplevda effekter samt hur bildkvaliteten påverkades enligt tekniska analyser.

Sammantaget bidrar resultaten till ökad kunskap om hur ultrahögfälts-MR påverkar både människan och undersökningsupplevelsen samt hur tekniska förbättringar kan användas för att göra undersökningarna mer skonsamma utan att kompromissa med bildkvaliteten. Detta är viktigt i takt med att 7 Tesla MR blir alltmer relevant både inom forskning och framtida klinisk diagnostik.

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