Acta Radiologica 37 (1996) 446–454
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ACTA RADIOLOGICA
ISSN 0284-1851

LOW-FIELD MR IMAGING – DEVELOPMENT IN FINLAND

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Abstract

The development project for application of MR imaging to diagnosis of internal hemorrhages was initiated by the Instrumentarium Corporation in 1978. The goal was to develop a diagnostic tool for emergency clinics. Due to the rapid development of imaging technology, the goal was changed to a cost-effective MR unit.

During the past 16 years, several generations of low-field units have been introduced. Consequently, a vast amount of clinical and technical knowledge about low-field MR has been gained.

The interest in low-field units is rapidly increasing. A part of this may be explained by the pressure to reduce the cost of health care. There are some features which make the low-field approach clinically interesting. These include the feasibility of open magnet configurations, and the availability of unique contrast parameters such as magnetization transfer and T1ρ. One important aspect is the inherent safety of a low-field MR unit.

This article reviews the methods and devices introduced through the development of low-field technology in Finland.

Key words: MR imaging, low field; methods, instrumentation; review.

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The development of low-field MR in Finland

The MR development in Finland was begun by the Instrumentarium Corporation in 1978. The original goal of the project was to develop a method and a device for detection of internal hemorrhages, as in emergency situations. During the first phase of the project, an experimental system operating at 0.17 T was constructed. The unit included a superconductive magnet which was produced in collaboration with the Low-Temperature Laboratory of Helsinki University of Technology.

The unit was transferred to the Helsinki University Central Hospital in June 1982 (Fig. 1). According to the original goal, the research concentrated on diagnosis of vascular disorders. Brain infarctions (24, 25) and hemorrhages (26, 27) were studied, especially by MR visualization of lesions at different times after onset of symptoms (24, 26).

Experience with the 0.17 T unit indicated that a different approach for an emergency MR unit had to be taken. Firstly, a reliable and characteristic visualization of an acute hemorrhage appeared to be rather difficult at 0.17 T (26). Secondly, the installation of a large and strong magnet in an emergency department seemed rather impractical.

The experimental results with an NMR spectrometer with a variable field strength indicated that a much lower field strength was needed. To confirm these results, 2 ultra-low-field (ULF) units were built, the first ULF unit operating at 0.01 T. Despite the low B_0, the resulting images were surprisingly good. The second experimental ULF unit operated at 0.02 T.

The expected visualization of a hematoma was confirmed with animal experiments at 0.01 and 0.02 T. For clinical trials, a 0.02 T unit was installed in the Emergency Department at Helsinki University Central Hospital (Fig. 2). The imaging results indicated that a ULF unit may be used for diagnosis of many pathologic conditions (20). Shortly afterward, the second unit was installed in the Radiology De-

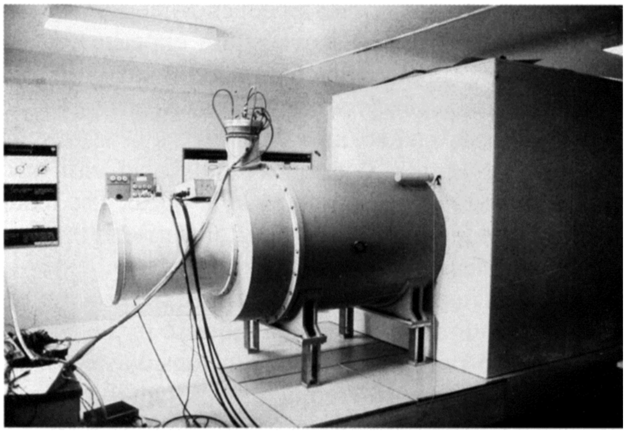

Fig. 1. The first experimental MR unit at Helsinki University Central Hospital was installed in June 1982. The 0.17 T superconductive magnet had a bore diameter of 60 cm. Over 3 000 patient examinations were performed during 1982–86.

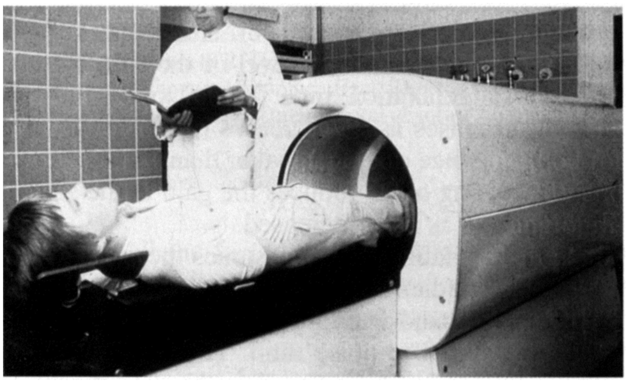

Fig. 2. The first 0.02 T unit was installed in the Emergency Department of Helsinki University Central Hospital in July 1983.

partment of Turku University Central Hospital. The first Acutscan commercial unit was installed in the Psychiatric Clinic of the Karolinska Institute in Stockholm in 1984. The unit was compact and rather easy to install in a standard hospital room. This was demonstrated by installation of a unit in the Emergency Department of New York Medical College (11).

As the applications of MR imaging expanded, the need for good spatial resolution, especially in spine and joint imaging, increased, and this led to the development of units with higher field strengths. The original idea of a scanner for emergency clinics was abandoned, and further development aimed at a cost-effective MR unit for imaging departments.

In 1988, the 0.04 T Magnaview MR unit was introduced. The next generation 0.1 T unit (MEGA 4) was introduced in 1990. After the Picker Nordstar Corporation was founded in 1993, the unit was sold as the Merit MRI unit. The Picker Nordstar Corporation was a joint venture of the Instrumentarium Corporation (Helsinki, Finland) and Picker International, Inc. (Cleveland, OH). Recently, Picker International, Inc. became the sole owner of the Picker Nordstar Corporation.

The low-field units mentioned above had a moderate commercial success. At the RSNA congress in 1994 in Chicago, the Picker Nordstar Corporation introduced the new Outlook system, which operates at 0.23 T. The iron-core resistive magnet has an open scanning area (Fig. 3). This feature allows examination of obese and claustrophobic patients and patients needing intensive monitoring. Semicinematic studies of joints, such as shoulders and knees, are feasible with this unit. It may even be possible to perform some interventional procedures during imaging. The Outlook MR unit has been well received on the market.

The collaboration between the Instrumentarium Corporation and the Hafslund-Nycomed Corporation began in 1988. This project is developing a novel diagnostic technique which exploits the Overhauser principle in the generation of high-contrast and high-resolution images. For this technique, a new class of contrast agents and a dedicated ULF scanner will be developed.

Since the beginning of the development project in 1978, much knowledge of MR techniques, clinical applications, and construction of cost-effective MR devices has been gained. In the following pages, some experience and results of the development work are summarized.

MR imaging and field strength

During the early phase of the clinical adoption of MR, the search for the optimal strength of the polar-

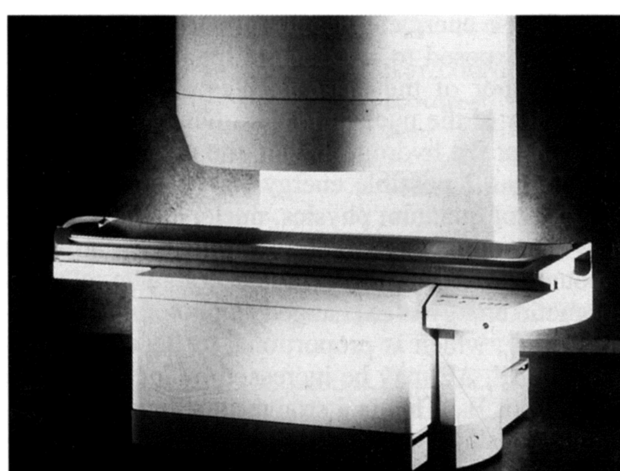

Fig. 3. The Outlook MR unit. The magnet is open and allows a good opportunity to monitor the patient during imaging.

izing magnetic field, B_0, was one of the major topics at scientific meetings and in journals. However, due to the diverse nature of the applications of MR, a universal optimum of the field strength has not been found. The continuous improvement of methods and devices has led to development of various MR units with field strengths of 0.02–3 T. It may be expected that this range will be even wider in the future.

MR units are generally classified into 3 categories by field strength: low-, medium-, and high-field units. This classification has only a weak theoretic basis and serves mainly commercial purposes. The borders between the categories are somewhat vague.

MR units with B_0 below 0.3 T have been considered low-field units. Specifically, the range below 0.1 T has been called the ULF range. Resistive or permanent magnet technology is used. The mid-field range is 0.3–1 T, most units are operating at 0.5 or 1 T. Most mid-field units include a superconductive magnet, but permanent magnet technology has also been used. Units with B_0 over 1 T are considered high-field units. Superconductive magnet technology is used. Scanners with B_0 ≥1.5 T may be used in spectroscopic studies and in scanning of nuclei other than protons.

Signal-to-noise ratio and field strength

The major drawback of the low-field MR unit is the low signal-to-noise ratio. However, this disadvantage is not as severe as might be expected from the basic NMR theory. The signal-to-noise ratio depends also on the properties of the object to be imaged and the scanning instrumentation.

MR imaging is based on the detection of electromagnetic radiation emitted by resonating atomic nuclei as they move from a higher to a lower energy state. These energy states are formed as soon as nuclei are exposed to an external magnetic field, B_0. The number of the energy states depends on the properties of the nucleus of a particular element. In the case of the hydrogen atom, the nucleus, i.e., the proton, has 2 possible energy states to occupy. By the laws of quantum physics, nuclei can change energy states only by absorbing or emitting an exact amount of electromagnetic energy. The electromagnetic energy is exchanged as photons at the frequency f_0, which is proportional to the energy difference ΔE. ΔE may be increased by increasing the strength of B_0. This is a straightforward way to improve the signal-to-noise ratio of an NMR experiment.

Theoretically, the signal-to-noise ratio is proportional to the square of B_0. When instrumental aspects are included, a weaker dependence may be expected. This is the case even with experiments involving small, nonconductive samples. As samples such as a human body are involved, the effect of the object must be taken into consideration. A human body is electrically conductive; hence, it will load the signal coil and generate electric noise. The loading effect increases as B_0, i.e., the resonance frequency f_0, increases.

As the size of the object increases, the loading effect rapidly increases. Hence, the gain in the signal-to-noise ratio of head and body images due to the increase of B_0 is considerably smaller than that of images of small parts; e.g., images of joints obtained with a small surface coil. Generally, when objects such as the human body are scanned, the maximal signal-to-noise ratio is proportional to B_0 (5).

Another important consequence of the increase of B_0 is the lengthening of longitudinal relaxation time (T1) of biologic tissues (7). This reduces signal due to a higher saturation level of the spin system. Transversal relaxation time (T2) also depends on B_0: T2 decreases as B_0 increases (4). However, the field dependence of T2 is weaker than that of T1 (2). The shortening of T2 reduces the gain in the signal-to-noise ratio as B_0 is increased.

Image formation in MR assumes the use of magnetic field gradients. The application of a gradient increases the band width of the signal and hence reduces the signal-to-noise ratio. The signal-to-noise ratio is inversely proportional to the square root of the band width. At high B_0, strong gradients are generally employed. This is to overcome the inhomogeneities of B_0 and the water/fat chemical shift misregistration effect (29). The inhomogeneity of B_0 is usually expressed in ppm:s, a measure which provides information about the relative inhomogeneity. However, for image formation, the meaningful information is the absolute amount of inhomogeneity which is expressed in the same unit (T) as B_0. A good absolute homogeneity is easier to reach at low-field than at high-field strength. This may be exploited in the generation of images with a high signal-to-noise ratio using a low band width of the signal or optimized sequences such as the completely balanced steady-state sequence (CBASS) (31). As an example of the high signal-to-noise ratio provided by the CBASS sequence, a neck image is presented in Fig. 4.

Losses and phase shifts of signals traveling through the body increase as B_0 increases (16). These phenomena are partially responsible for artifactual intensity variations and ghosting due to motion. These artifacts plague more high-field than low-field images.

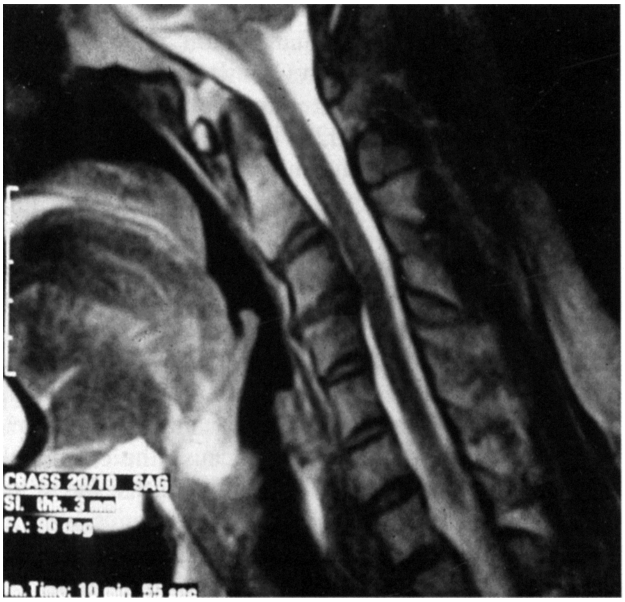

Fig. 4. A CBASS image of a neck obtained at 0.1 T. The sequence parameters (TR/TE 20/10 ms, α 90°) were optimized for generation of the myelographic effect. The completely balanced steady-state sequences assume a good absolute homogeneity of the B_0 field.

When all the factors mentioned above are taken into consideration (Fig. 5), one could expect the signal-to-noise ratio to increase more slowly, in proportion, than B_0. A significantly weaker dependence may be expected for large objects scanned with a large-volume coil (e.g., head, abdomen, pelvis) and a stronger dependence for small objects scanned with a small-surface coil (e.g., temporomandibular joint, eye). The exact prediction of the signal-to-noise ratio on the strength of B_0 is difficult due to the heterogeneity of the composition of the human body.

Tissue contrast and field strength

Longitudinal relaxation times (T1s) of biologic tissues are dependent on B_0. At low-field strengths, the relaxivity of macromolecule/water interaction is more efficient than at higher field strengths. The magnetization transfer rate between protons of macromolecules and protons of water molecules increases as B_0 decreases (8). It has been suggested that T1 relaxation at very low magnetic fields is mostly due to the magnetization transfer (3). Generally, the T1 of biologic tissue is proportional to the cubic root of B_0. T1s of liver and muscle tissue have stronger dependence on B_0 than most other tissues. T1 of fat has only a weak dependence. T1 of water does not demonstrate a significant field dependence in the range of B_0 used in clinical MR imaging (7).

One could expect that T1 contrast between some tissues improves as B_0 decreases. For example, this is the case for tissue contrast between the spleen and the liver. However, T1 contrast between grey and white matter decreases as B_0 decreases. Because the T1 of water is not field dependent, the T1 contrast between lesions with an increased water content and adjacent normal tissue will be improved as B_0 decreases. Low-field MR imaging has demonstrated good diagnostic capability in detection of hematomas and infections (6). As an example, the visualization of a liver abscess with a T2-weighted image at 0.02 T is presented in Fig. 6. There are some indications that an ultra-low magnetic field (0.02 T) may provide better visualization and characterization of intracerebral and subdural hematomas than a low field (0.17 T) MR (28). It has also been suggested that a low field provides better correlation between 1/T1 and brain tumor tissue than a high field (12).

A shorter T1 and a longer T2 are advantageous

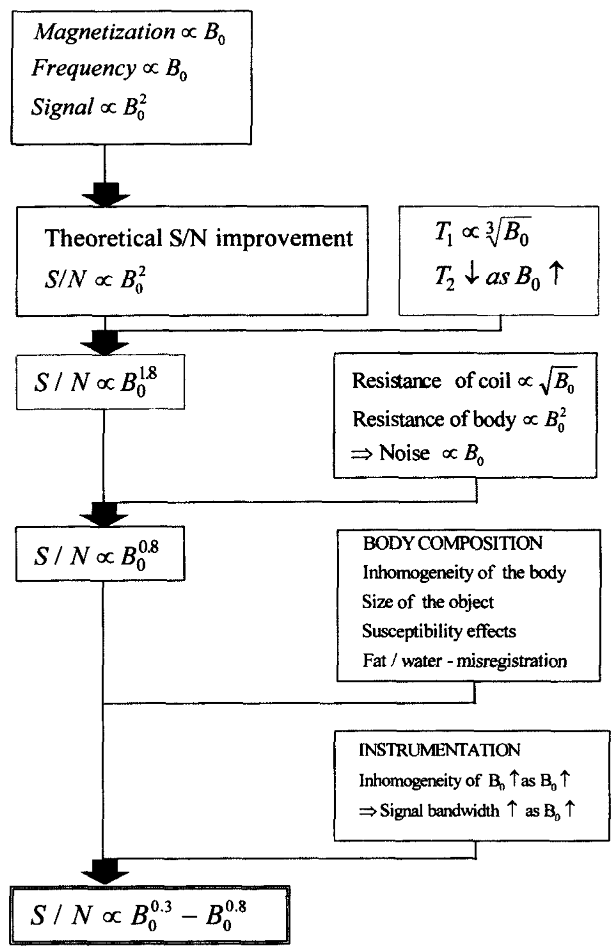

Fig. 5. A summary of the factors affecting the relationship between the signal-to-noise ratio and the strength of B_0.

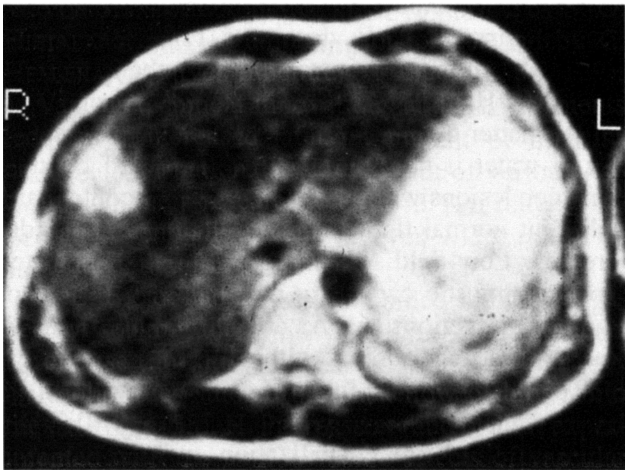

Fig. 6. A liver abscess. The scanning sequence is T2-weighted SE 2000/100. The strength of B_0 is 0.02 T.

for the contrast of the T2- and proton density-weighted images because of the more complete recovery of the spin system during the repetition time (TR). Therefore, one may expect some improvement of the T2 contrast as B_0 is decreased.

The contrast-to-noise ratio of the final image is the essential parameter in assessment of the diagnostic capability of any imaging technique. Because the signal-to-noise ratio decreases as B_0 decreases, the improvement in the inherent tissue contrast usually reduces the loss in the contrast-to-noise ratio. It may be concluded that the contrast-to-noise ratio generally demonstrates even smaller B_0 dependency than the signal-to-noise ratio. The loss may be reduced or even compensated by using signal averaging or a field-cycling technique (15).

Magnetization transfer and spin-lock

Magnetization transfer (MT) is a recently introduced contrast parameter in MR imaging. MT is mainly due to the cross-relaxation between the pool of mobile protons H_f and the pool of protons with restricted mobility H_r. H_r includes protons of macromolecules and bound water molecules. H_f consists of protons of freely moving water molecules.

In order to generate an MT contrast, one saturates the magnetization of protons of macromolecules, i.e., of H_r. The saturation is then transferred via cross-relaxation to the magnetization of protons of mobile water molecules, i.e., of H_f. The more efficient the interaction between water molecules and macromolecules, the more significant is the decrease of the magnetization of mobile protons (34).

MT contrast improves the visualization of lesions with increased water or fat concentration. The first demonstrations of a human lesion with MT was done with a low-field unit (30). It may also be used to improve the contrast enhancement with paramagnetic substances (32). To demonstrate the MT contrast between fat and muscle tissue, a leg image is presented in Fig. 7.

In a spin-lock experiment, the nuclear magnetization is locked along the magnetic component of the radio frequency (RF) field B_{1L}. The magnetization relaxes toward an equilibrium value which is approximately zero (19). The relaxation under the locking condition seems to be mainly due to the magnetization transfer between water protons and protons of macromolecules. Because the efficiency of MT is high at very low-field strengths, the spin-lock technique may be the most efficient way to generate an MT-based contrast in protein-rich tissues (3).

By varying the strength of B_{1L}, one may generate images based on the tissue-dependent variation of $T1\rho$; i.e., $T1\rho$-dispersion. This unique tissue information has proven to be useful in diagnosis of disorders of the brain (22) and muscle (9).

In both techniques, one applies an additional RF irradiation pulse for contrast generation. This limits their applicability at high field strengths.

Further development of sequences with short locking or saturation pulses and low flip angle excitations has made it possible to utilize these techniques at medium and high field strengths (17, 18).

Fig. 7. An MT-weighted image of a leg obtained with a gradient echo sequence at 0.1 T. The scanning parameters are TR/TE 100/20 ms, α 50°. The off-resonance pulse parameters are B_1^{off} 30 μT, TS 50 ms, frequency offset 4 000 Hz.

Chemical shift imaging

Chemical shift imaging is used to distinguish between fat and water components of tissues and may be used in the diagnosis of bone-marrow (21) and muscle (10) diseases. This technique utilizes the 3.4 ppm difference between the resonance frequencies of protons of water and fat molecules. At 0.02 T, the frequency difference between resonating water and fat protons is only 3 Hz. Therefore methods which require large phase differences generated by the chemical shift cannot be used. However, by using phase information, chemical shift imaging may be performed successfully even at ULF strengths (33).

Chemical shift information may be derived from a conventional gradient echo image if the phase information is stored. At low field, the susceptibility and other phase-distorting effects of the human body are negligible. This enables one to use simple techniques for correction of magnetic field inhomogeneities. A water phantom may be used to map the inhomogeneity of B_0. This map is then used for correction of the phase information obtained from the object to be scanned. The needed phase difference between the water and fat signals is relatively small. At 0.02 T, a gradient echo with a TE of 40 ms has been used for generation of fat and water images. The benefit of this technique is that it does not increase the scanning time. However, this technique is difficult to use at high fields due to the susceptibility variations and phase-distortion effects of the human body.

Practical considerations

Safety

A magnetic field exerts magnetomechanical forces on ferromagnetic objects. This may be the most serious health risk of MR imaging. The magnetomechanical force is dependent on the strength of B_0.

The forces may be strong enough to cause even heavy objects to fly toward the magnet (the missile effect). A strong magnetic field may also be able to cause a movement of metallic implants or other ferromagnetic objects in the body. At low field strengths, the potential risks of missile effects and forces on ferromagnetic implants are considerably less serious than at high field strengths (23).

The specific absorption ratio (SAR) is proportional to the square of the product B_0B_1 (1) where B_1 is the excitatory RF field. Because the electric losses in the body increase as the frequency increases, the RF power needed for RF operations, e.g., for excitation pulses, increases. The RF power is proportional to the square of B_1. The RF power may cause injuries by local heating if the patient is accidentally coupled in the RF circuit via loops of the cables of the signal coil. The ECG and pulse oximetry monitors may also be hazardous if the RF power leaks through electrodes or transducers. At low field strengths, the irradiation power is much smaller. Additionally, stray couplings of the power are very weak due to the low frequency. Therefore it may be concluded that RF irradiation during low-field MR imaging involves much smaller risks than during high-field imaging.

Acoustic noise is a less serious problem at low field strengths than at higher field strengths. At 0.02 T, the gradient operations are nearly inaudible. At 0.2 T, the noise is clearly audible but still very tolerable. A low noise level is beneficial in imaging of claustrophobic or anaesthetized patients. The low level of acoustic noise is also important during eventual interventional operations performed under MR guidance.

Magnet design

Medium- and high-field magnets utilize superconductor technology. Superconductive magnets require refilling with cryogens. Special systems must be installed in order to fulfill the safety requirements. A low-field magnet is easier to install in a hospital environment because the fringe fields are more easily eliminated than those of a higher field magnet. The safety requirements are not so demanding as with high-field units.

It is obvious that a low field strength is easier to generate than a high field strength. This benefit gives a certain freedom to design the magnet from other viewpoints than just the maximal efficiency in generation of the magnetic field. Magnets with a solenoid structure are most often used in MR units. This structure can generate a homogeneous magnetic field with a reasonable current. The drawback is the tubular shape of the magnet. These scanners are known to cause anxiety and claustrophobic reactions in some patients. This structure seems not to be suitable for units used in interventional studies.

Recently, several low-field units with an open magnet structure have been introduced. The C-structure has certain advantages: The iron core enables effective generation of the magnetic field. Either resistive or permanent magnet technology may be used. The C-structure offers good access to the patient. The openness reduces claustrophobic reactions and allows good monitoring of the patient.

These units are suitable for some interventional studies.

Future developments

"Low-cost" MR units

It has been speculated that very "low-cost" MR units may be devised by the field-cycling technique. In this method, the polarization of nuclear spins takes place at higher field strength than for the detection of resonance signals. The prepolarization field does not need to be as homogeneous as the detection field. This reduces considerably the requirements for the magnet construction. A pulsed MR system has already been used to generate T1 dispersion images. The predicted improvement of signal-to-noise ratio has also been demonstrated. This approach may yield a new generation of "low-cost" MR scanners for specific applications (15).

Overhauser enhanced MR imaging (OMRI)

The application of the Overhauser technique assumes the existence of 2 interacting spin populations. In the OMRI technique, one transfers the magnetization of electron spins to proton spins. This transfer is enabled by saturating the magnetization of electron spins. The saturation is transferred by the dipolar relaxation process to the nuclear spins. This leads to a considerable increase of the magnitude of the magnetization of nuclear spins. In the case of protons, one may have an up to 330-fold increase of magnetization. In practice, one may expect an increase by a factor of 50–100. This leads to a corresponding increase in the signal-to-noise ratio.

An application of the method assumes an introduction of a paramagnetic substance into the body. Therefore, a special contrast agent must be developed. In order to reduce the electron spin resonance (ESR) irradiation power and have a maximal gain from the technique, the ESR spectrum of the contrast agent must be as narrow as possible. For example, conventional gadolinium-based substances are not applicable.

The OMRI technique seems to be applicable only with ULFs. This is because the ESR frequency is about 660 times higher than the corresponding NMR frequency. It seems that in studies of the human body the field strength during the ESR irradiation should be less than 0.01 T (13). However, by using the field-cycling technique, a stronger detection field may be employed (14).

Therefore, one may expect a signal-to-noise ratio which corresponds to a B_0 of 0.5–1 T and a very high contrast between tissues with different enhancement levels. In OMRI images, only tissues with contrast agent are visible (Fig. 8). The method complements conventional imaging methods and may find application in angiography, perfusion studies, and other functional imaging experiments.

Fig. 8. An OMRI image of a rat obtained at 0.01 T with the gradient echo technique. The scanning parameters are TR/TE 300/20 ms; the ESR pulse length for saturation of the electron spin system is 250 ms. Only tissues with a paramagnetic contrast agent are visible. (The image was provided by Dr. G. Ehnholm.)

Conclusions

Low-field MR imaging seems to have gained wider acceptance in the clinical environment. The practical benefits of low-field MR are safety, low cost, and easy installation in a hospital. These facts, in view of the pressure to reduce the cost of health care, can explain the growth of the interest in low-field MR units.

However, there are also some diagnostic potentials, which may give a unique advantage to low-field MR. The freedom to design a magnet with an open structure enables studies of claustrophobic and obese patients. This feature may be used also in cinematic studies of joints and in interventional procedures.

Tissue-specific information may be obtained by MT and T1ρ imaging. Relaxometric information which may prove to be more tissue specific than the conventional relaxation time information may be obtained by using T1ρ dispersion or the field-cycling technique.

Application of the field-cycling technique may provide a new approach to the design of "low-cost" MR units and even to the collection of relaxometric information which may be used in tissue characterization.

The use of the Overhauser principle for enhancement of the proton resonance signal will open a new field of MR diagnostics. The inherently high contrast and signal-to-noise ratio will provide a new tool for studies of the function of the human body. In the future, the OMRI technique may be exploited in studies of tissue perfusion and other important features. Hence, one may expect that low-field MR will find more applications in diagnostic and interventional radiology.

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