Magnetic resonance imaging in canine spontaneous neurological disorders: an evaluation of equipment and methods

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ACADEMIC DISSERTATION

To be presented, with the permission of the Faculty of Veterinary Medicine, for public criticism in Auditorium Maximum, Hämeentie 57, 00580 Helsinki on the 13th of June 2000 at 12 o’clock.
“With MRI, neuroradiology becomes gross pathology, in vivo, when knowledge of the pathology may still be helpful to the patient.”


To Eero and Jyri
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LIST OF ORIGINAL PUBLICATIONS

The thesis is based on the following publications, which are referred to in the text by Roman numerals. In addition, unpublished data are presented.


The publishers have kindly permitted reprinting of the original articles.

In the following text the original reports are referred to by their Roman numerals.
ABSTRACT

Ultralow-, low- and high field strength magnetic resonance units were used for test imaging of ten normal dogs’ brain and spinal cord and to diagnose spontaneous brain and spine diseases in 56 dogs. The devices were designed for human use and suitable procedures for immobilization, positioning, and imaging sequence protocols for canine examinations had to be created and tested. Normal tissue characteristics with different types of scanners and sequences had to be established for interpretation of the magnetic resonance images findings in patients.

Immobilization for scanning by sedation with medetomidine alone or combined with methadone proved to be efficient, practical and safe for dogs. Each dog was positioned individually in sternal and lateral recumbency to adjust for the anatomy and the type of the local coil. The standard human coils that were best suited for brain MRI were the knee coil, and for spine, the spinal coils. In low field imaging, flexible multipurpose local coils could also be used for brain and lumbar spine scanning. A dedicated brain coil for Beagle-sized dogs was made for the low field strength scanner. It improved quality of brain images compared with the standard coils with an improved signal-to-noise-ratio (SNR) of more than 20%, as compared with that of the flexible coil, and 70%, as compared with the standard knee coil supplied by the manufacturer.

Routine T1 and T2 weighted scans were made in sagittal and transverse planes. The dorsal plane was used if more detailed information of a lesion was needed. After contrast application T1 weighted scans were made. Proton density weighted scans did not add to the diagnosis. The shortest scanning protocol was used in severely ill patients where T1 or T2 weighted transverse-oblique scans could show brain tumours. The T2 weighted scans could show oedema around these tumours well.

The imaging sequences were designed individually or if a similar dog, for example, a littermate was scanned the same sequences could be used. For scanning the same dog in different field strength machines different sequences had to be used. Magnetic resonance imaging requires good cooperation between the hospital physicist and the veterinarian to produce the best possible sequences for the particular machine to use on dogs. This work is as necessary an investment as the equipment. The sequences selected for use on each patient from those available must be chosen with regard to the suspected lesion. An asymmetry of the lateral ventricles was seen frequently in patients but also in one normal dog. Other investigators have also seen variability in size and symmetry of the lateral ventricles of normal dogs. It can be concluded that a variation like this is not necessarily a pathologic finding.

Contrast enhancement with Gd-DTPA 0.1-0.2 mmol/kg iv helped to show the lesions in brain in patients with astrocytoma and borreliosis. In the patients in which no enhancement was seen it helped to rule out conditions with ruptured blood-brain-barrier. To save imaging time in patients with very poor condition and in which lesion was seen in T1W and T2W scans, no contrast was used. No adverse reactions for contrast medium were observed.

The scanning of the whole spinal cord requires at least three different sagittal planes because of the curving of spine. If the spinal cord is not positioned exactly straight for sagittal planes artifacts will make the evaluation difficult or impossible.

In low field strength spinal cord imaging the sagittal planes show compressions and disks much better than transverse ones because of the small size of the spinal cord and the poorer resolution in high fields. High field strength scans of spinal cord could show details of lesions as well in transverse as in dorsal scans because of the improved resolution compared with low field strength scans. The detail of a high field machine is necessary if surgery of spinal tumours is planned.

The low field strength scans gave satisfactory images of canine brain anatomy and disorders. However, the lesions of encephalomalacia could not be detected because of the low resolution of an ultralow field machine.

In the dogs that were scanned also in high fields the latter did not add to the findings of hydrocephalus, meningomyelocele, and brainabscess.

The ultralow-, low-, and high field magnetic resonance scanners used in this investigation could all be used for diagnosis of central nervous disorders in dogs. MRI proved to be a practical method for veterinary use.
## ABBREVIATIONS

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<th>Definition</th>
<th>Symbol</th>
<th>Definition</th>
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<td>B&lt;sub&gt;o&lt;/sub&gt;</td>
<td>Constant magnetic induction field in a NMR system</td>
<td>R&lt;sub&gt;2&lt;/sub&gt;</td>
<td>Transverse relaxation rate (R&lt;sub&gt;2&lt;/sub&gt;=1/T&lt;sub&gt;2&lt;/sub&gt;)</td>
</tr>
<tr>
<td>BBB</td>
<td>Blood-brain-barrier</td>
<td>ROI</td>
<td>Region-of-interest</td>
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<td>C</td>
<td>Cervical vertebra</td>
<td>SE</td>
<td>Spin echo</td>
</tr>
<tr>
<td>CBASS</td>
<td>Completely balanced steady state</td>
<td>SL</td>
<td>Spin lock</td>
</tr>
<tr>
<td>CNS</td>
<td>Central nervous system</td>
<td>SNR</td>
<td>Signal-to-noise-ratio</td>
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<tr>
<td>CSF</td>
<td>Cerebrospinal fluid</td>
<td>SR</td>
<td>Saturation recovery</td>
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<tr>
<td>CT</td>
<td>Computerized tomography</td>
<td>T</td>
<td>Tesla</td>
</tr>
<tr>
<td>2-D</td>
<td>Two dimensional</td>
<td>Th</td>
<td>Thoracic vertebra</td>
</tr>
<tr>
<td>3-D</td>
<td>Three dimensional</td>
<td>T&lt;sub&gt;1&lt;/sub&gt;</td>
<td>Longitudinal relaxation time</td>
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<td>DE</td>
<td>Dual echo</td>
<td>T&lt;sub&gt;2&lt;/sub&gt;*</td>
<td>T-two-star</td>
</tr>
<tr>
<td>FA</td>
<td>Flip angle</td>
<td>TE</td>
<td>Time-to-echo</td>
</tr>
<tr>
<td>FAST</td>
<td>Fourier acquired steady state</td>
<td>TI</td>
<td>Inversion time</td>
</tr>
<tr>
<td>FISP</td>
<td>Fast imaging with steady state precession</td>
<td>True-FISP</td>
<td>Fast imaging with steady precession (heavily T&lt;sub&gt;2&lt;/sub&gt;* weighed)</td>
</tr>
<tr>
<td>FLASH</td>
<td>Fast Low Angle Shot imaging</td>
<td>TR</td>
<td>Repetition time</td>
</tr>
<tr>
<td>FOV</td>
<td>Field-of-view</td>
<td>T&lt;sub&gt;1&lt;/sub&gt;W</td>
<td>T1-weighted</td>
</tr>
<tr>
<td>FSE</td>
<td>Fast spin echo</td>
<td>T&lt;sub&gt;2&lt;/sub&gt;W</td>
<td>T2-weighted</td>
</tr>
<tr>
<td>G</td>
<td>Gauss</td>
<td>WH</td>
<td>Wire-haired</td>
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<tr>
<td>Gd-DTPA</td>
<td>Gadolinium-diethylene-triaminepentaacetae</td>
<td>(\alpha)</td>
<td>Flip angle</td>
</tr>
<tr>
<td>GRE</td>
<td>Gradient echo</td>
<td>(\gamma)</td>
<td>Gyromagnetic ratio</td>
</tr>
<tr>
<td>IR</td>
<td>Inversion recovery</td>
<td>(\nu)</td>
<td>Frequency</td>
</tr>
<tr>
<td>L</td>
<td>Lumbar vertebra</td>
<td>(\tau)</td>
<td>Time delay</td>
</tr>
<tr>
<td>LS</td>
<td>Lumbo-sacral</td>
<td>(\rho)</td>
<td>Proton density</td>
</tr>
<tr>
<td>Im</td>
<td>Littermate</td>
<td>(\chi)</td>
<td>Susceptibility</td>
</tr>
<tr>
<td>MRI</td>
<td>Magnetic resonance imaging</td>
<td>(\omega)</td>
<td>Larmor frequency is the frequency of oscillation or rotation of the protons in a given magnetic field</td>
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<tr>
<td>NMR</td>
<td>Nuclear magnetic resonance</td>
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<td>PDW</td>
<td>Proton density weighted</td>
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<tr>
<td>PS</td>
<td>Partial saturation</td>
<td></td>
<td></td>
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<td>RF</td>
<td>Radio frequency</td>
<td></td>
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<tr>
<td>R&lt;sub&gt;1&lt;/sub&gt;</td>
<td>Longitudinal relaxation rate (R&lt;sub&gt;1&lt;/sub&gt;=1/T&lt;sub&gt;1&lt;/sub&gt;)</td>
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INTRODUCTION

The imaging of the diseases of the canine central nervous system with methods of radiography, computed tomography (CT) and diagnostic ultrasound gives only limited information about the pathologic-anatomic changes in the tissues. Ultrasound cannot penetrate the bone protecting the brain and spinal cord. Radiography cannot distinguish central nervous system (CNS) tissue from other soft tissues. CT produces cross sectional images, which discriminate CNS but with little contrast of the CNS tissues. The increasing skills in clinical diagnosis and surgery of CNS diseases have created a demand from the veterinarians for a reliable prognosis. The humane aspect and the animal welfare viewpoint must also be taken in consideration. The owners feel great anxiety when their pets suffer from neurological problems such as seizures, paralysis or changes in behaviour. They want quick and reliable answers about the diagnosis, the cause, and possible therapy. Neither the owners nor the veterinarians nowadays accept euthanasia without a well defined reason, yet the prognosis of a neurological ill canine patient can be difficult to determine from a clinical examination.

Magnetic resonance imaging (MRI) offers a method that gives cross sectional images and high contrast of the tissues comprising the CNS. However, the method is far more complex in its physics and the application of the principles than is any of the other imaging methods. At the time of the beginning of these studies, little was known about the optimal MRI techniques in human patients, and even less in dogs. The first atlas on canine MRI was not published until 1997 (Assheuer et. al).

Since the initial experiences with clinical MRI in human medicine (Holland et al, 1980a, 1980b; Young et al,1981; Bydder et al,1982), MRI has proven to be an especially effective technique in neurology. MR images of the brain enable a good distinction of anatomic details because of the different relaxation times of white and grey matter both in man (Doyle et al, 1981) and dogs (Armstrong 1983) and allow anatomical discrimination of these tissues. The high cost and limited availability of MR scanners has been restricting its use in veterinary medicine. The low field strength scanners are more economical to buy and maintain than the high field units (Bailey 1990) and therefore their potential in veterinary diagnostic imaging is of special interest. Before the beginning of our MR studies in 1987 on canine neurological patients with ultralow- and low field strength units such imagers had been used only by Armstrong et al, (1983) with a 0.15T unit. Salvatore et al, (1987) used a mid-field 0.5T unit for the diagnosis of spontaneous diseases of the canine brain. In 1987 the development of a Finnish made ultralow field 0.02T scanner gave me an opportunity to apply this method to canine spontaneous neurological diseases with help of a researcher developing these devices. The machine was replaced by a more powerful 0.04T unit in 1989. In 1991 a 0.1T low field strength imager was ready to be used. From 1990-98 a number of healthy control and patient dogs were scanned in one of the two high field strength 1.0T scanners. All these MRI devices were made for human patients and their local coils were not designed for canine anatomy. We therefore designed a brain coil for use with a low field unit. The value of these scanners and their most effective application in dogs needed to be determined.

HISTORICAL BACKGROUND OF MRI

Felix Bloch and Edward Purcell in the USA independently discovered in 1946 the physical nuclear magnetic resonance (NMR) phenomenon. For about twenty years NMR served solely the research of chemistry, biology and physical composition of materials (NMR spectroscopy). Damadian observed in 1971 in vitro that the longitudinal relaxation times of hydrogen in cancer tissue and normal tissue differ from each other and suggested that spin echo nuclear magnetic resonance measurements may be used for this discrimination. In 1973 Lauterbur suggested that the principles of magnetic resonance (MR) should
be used for imaging purposes. He called this technique zeugmatography (zeugma = the yoke, “two magnet fields were to be joined”). Mansfield et al, (1974) published the first image of a living organ – a finger. After Lauterbur several other NMR research groups developed their own basic imaging techniques (Garroway et al, 1973;1974; Mansfield et al, 1975; Mansfield et al, 1977; Mansfield 1977).

In 1977 Hinshaw et al, made the first NMR image of the human wrist. The first high quality magnetic resonance imaging (MRI) scans of healthy and diseased human brains appeared in 1980 (Holland et al,1980a;1980b; Hawkes et al,1980). Low field scanners were early found suitable for brain imaging in humans. Doyle et al, (1981) observed with a 0.15T scanner the striking differentiation between grey and white matter within the brain based on tissue-specific relaxation times. Agartz et al, (1987) found the 0.02T scanner to have a high contrast resolution. High contrast contributes to the detection of pathological changes and metabolic alterations in tissue. The word “nuclear” was left out later from the description of MRI as the word may have negative connections for the public.

The application of MRI in dogs began because of research in human neurology (Brant-Zawadzki, 1984; Ngo et al, 1985; Runge et al, 1985a), 1985b); Chakreres et al, 1987; Whelan, 1987). The development and testing of MRI contrast media was based from the very beginning on animal experiments, in which dogs were often used (Bydder et al, 1982; Brasch et al, 1983; 1984).

In neuroradiology in human medicine in recent years MRI has become the examination of choice (Caillé 1995).

**BASIC PHYSICS**

Hydrogen atoms are by far the most common atoms in living tissues, bound in different ways in the compounds which comprise the tissues. In water, the hydrogen nucleus (proton) is influenced by electrons in H-O bonds and in fat it is influenced by electrons in H-C bonds (Rinck 1993). MRI is based on mapping the density and magnetic properties of hydrogen atoms in locally varying magnetic field in the tissue or organ being examined (Sepponen et al, 1984). The different degrees of binding affect the sensitivity to the applied magnetic field and thus the appearance of different tissues. MR images of biologic tissues are based on signals emitted by the pool of freely moving water molecules. There is also a pool of protons in macromolecules and water molecules with restricted mobility (Sepponen et al, 1984). The relaxation rate of these spins is too high to be observed by conventional imaging methods but saturating them affects the relaxation rate of the free spins’ pool, which can be used to obtain contrast in MRI images.

**THE COMPONENTS OF THE EQUIPMENT**

The basic components of an MR-imager are a magnet, gradient coils, radiofrequency-pulse transmitter and receiver coils, and a data acquisition system including computers, power suppliers and cooling systems (Rinck 1993). A powerful magnet creates a homogenous static magnetic field inside the bore of the device. The patient couch supports the RF coils and the patient. The patient is moved to inside the bore for imaging so that the part to be imaged comes into the middle of the magnetic field. RF coils are used to send an excitation pulse and receive the signals from the tissue. Gradient coils that create magnetic fields of varying strength and intensity along and across the body being imaged, are a requirement for spatial localization of the MR signals from the body. The reconstruction of an image is made with the data acquisition system with computers from the information of the collected signals. The magnet has to be cooled: with liquid nitrogen or helium to preserve the superconductivity of the magnetic coils in the high field strength units. Only water cooling is necessary in low field units.
Theory

The main magnet produces a very strong and ideally uniform external magnetic field. Protons have a positive charge and spin, and thus the patient’s tissues can be magnetized by the strong magnetic field. The strong external field aligns these small magnets in the field from their previously random and neutral arrangement. The alignment may be in one direction with the field or the opposite. There are slightly more protons aligned with the magnetic field, because this group has a slightly lower energy level. So the result is a very weak magnetization in the patient. The protons, when aligned, have a movement like that of a wobbling spinning top. The movement of the axis of the spin is in the form of a cone. The center of the cone is aligned with the magnetic field. The greater the strength of the magnetic field, the faster the rotation. This relationship is described by the Larmor equation:

$$\omega_0 = \gamma \cdot B_0$$

where $B_0$ is the magnetic flux density, $\omega_0$ is the Larmor frequency, and $\gamma$ is the gyromagnetic ratio which numerical value for proton is about 42.58 MHz/T. It is a common practice in the literature of MRI to use $B_0$ to symbolize also the magnetic field strength, which notation is used in this work as well.

Equipment

The field strengths of most clinically used MR-imagers range from 0.02T to 3.0T (1 Tesla = 10 000 Gauss). The earth’s magnetic field (50 mT, 0.5 Gauss) is only a very small fraction of this. The main magnetic field can be produced by a permanent, resistive or superconducting magnet. For clinical use it is possible to produce a magnetic field up to 0.3T with a resistive or permanent magnet. MR-scanners operating at 0.5T or more use superconducting magnets, which must be cooled with liquid helium to allow superconducting to occur. These may cause two potential dangers if the cooling gases leak to the examination room: frostbite and oxygen substitution causing breathing difficulties. Constructional methods can eliminate these risks by leading the gases out of the examination room.

In this investigation the MR-devices are classified according to field strength (Sairaalalitto, The Finnish Hospital Association 1991):
ultralow field, under 0.1T,
low field, under 0.3T,
mid field, over 0.3T to 0.99T, and
high field strength imagers, over 1.0T.

The earlier publications of this thesis used a different classification, which explains the use of “low field” for what is now called “ultralow field”.

The magnetic field ($B_0$) should be homogenous to give the best field characteristics so that the dephasing and signal decay is dependent only on the properties of the material and its compounds. Absolute homogeneity of the magnetic field is important because it is a requirement for effective imaging with gradient echo sequences (Virolainen et al, 1993). MRI machines are delivered with shim coils for reducing field inhomogeneities. When currents are passed through these coils, correctional fields of known geometry are produced to compensate for the inherent inhomogeneity of the magnet (Rinck 1993). They are tuned and tested with phantoms when the machine is installed.

Radiofrequency coils

Theory

The magnetization of the patient cannot be measured, and even if it could, it would not show differences between tissues. The aligned protons
have to be disturbed to make them send out their own signal. This is done by sending into the body radio waves with a radiofrequency (RF) pulse of the same frequency as the precessing protons, that is, their Larmor frequency, which can be calculated. When these energies are the same, energy from the radio waves can be transferred to the protons. When all the protons are made to “flip” at the same time they have coherent resonance and are given higher energy by the RF pulse. This new energy causes the protons to do two things. The spins precess in phase with respect to the excitation RF field and the net magnetization will in general have a component which is orthogonal to the static magnetic field. This precessing magnetization can be detected by an external antenna. After the RF pulse is stopped the protons lose this energy. This energy released also matches exactly the difference between the two energy levels for protons dictated by the magnetic field strength and can be detected with the antenna (in the coil) around the region of interest (Brant-Zawadzki 1987).

Radiofrequency (RF) coils are used to transmit the RF pulses, and to receive the MR signal. Transmitting coils send precisely placed, timed and tuned RF pulses into the tissues, which set the spinning nuclei in phase. For effective reception and transmission, the oscillating RF magnetic field of the RF coil must be perpendicular to the main magnetic field Bo.

Different types of RF coils are presented in FIG. 2.

A whole body coil is permanently inside the bore, and works both as the transmitting and receiving coil. A head coil is a special receiver coil having close contact with the human head to improve the reception.

A variety of coils have been designed to improve the attainable signal-to-noise ratio (SNR). The form of the coil determines upon the homogeneity of the radiofrequency field. Usually solenoid and saddle coils are used. A linear coil transmits RF energy into the patient with linearly polarized magnetic field which can be described as two fields which rotate in opposite directions, and only one rotational component can be used for signal excitation and the other only generates heat in tissue. A modification to the standard linear technology is the quadrature excitation and signal detection, where two RF channels are used and the second channel is phase shifted by 90° with respect to the reference used for the first channel (Rinck 1993). In the transmit mode this reduces the RF energy absorption by a factor of two, and in signal detection it improves the SNR by about 40% compared to the linear detection. (Brant-Zawadzki 1987).

Surface coils

The advantage of using surface coils is that the small volume of tissue close to the coil creates a better signal-to-noise ratio (SNR) (Kneeland et al, 1989; Rinck 1993). The better the object fills the coil, the better the image quality. The use of local coils to increase the resolution of MRI has considerably enhanced this modality’s capability (Kneeland et al, 1989). Small surface receiver coils are closely applied to the anatomic region of interest (Osborn et al, 1992) and image only a small region of anatomy. The surface coils do not produce as homogenous RF field as other coils. The term “surface” coil has been applied to describe this class of coils that partially or completely surround the structure of interest. This term may be misleading and therefore Kneeland et al, (1989) prefer to use the term “surface” coil for the subset of this class of coils consisting of flat loops. The SNR is high as it receives little noise from that small volume. Signals from outside the coil’s sensitive volume (breathing movements

FIG. 2. Different types of coils for 0.1T scanner. A) Head coil. B) Body coil. C) Spinal coil. D) Knee coil (Picker Nordstar Inc.).
for example) cannot cause imaging artifacts. Surface coil MRI allows thin sections with high spatial resolution, which reduces partial volume averaging (Pauschert et al, 1985). Surface coils that are made to image small superficial areas lose their signal intensity rapidly in deeper (greater than 5-6 cm) tissue layers. The MR signals are detected from a small region of interest as close to this coil as possible when it is placed, for instance, on a patient’s head. In imaging of dogs mostly human local coils like head, whole body (Kraft et al, 1989), knee (Hatchcock 1996) and spinal coils have been used as no specific coils for dog imaging were available.

Low field strength (0.1 - 0.3T) MR units can provide satisfactory anatomic details of the canine brain (Sepponen et al, 1985). The image quality of a low field strength imager can be enhanced using a dedicated coil for signal detection. A surface coil that surrounds the dog’s head with a minimal possible diameter gives the optimal depth penetration for signals and thus increases SNR. The improvement in SNR can either be used to obtain images of a better quality or to reduce scanning time.

Gradient coils

Theory

The MR signals from the body have to be localized to create the three dimensional picture of the body. The purpose of the gradient coils is to create differences in the magnetic field, a gradient, in the three planes of the body that can be used for spatial localization of the MR signals. Each part of the body thus will have a different field strength, and by the Larmor equation, a different emitting frequency.

Equipment

Gradient coils are electromagnetic coils built within the magnet. Gradient coils are typically wound as a single assembly that contains coils for all directions and fit in the magnetic bore inside the shim coils (Brant-Zawadzki 1987). They produce fields much weaker (approximately 100 times lower in mid- and high field systems) than the main magnetic field. This field creates a specific desired gradient (gradient = the amount and direction of the rate of change in space of some quantity, such as magnetic field strength, Rinck 1993) in three planes or directions (x, y and z). The strength of these gradient fields varies linearly so that each spatial location in the tissues has a unique magnetic field and unique Larmor frequency. The z-gradient defines the slice thickness. The phase encoding gradient defines signals on the y-plane of the image and the frequency encoding gradient (readout gradient) on the x-plane. In some systems the direction of the phase encoding and the frequency encoding gradients can be changed.

When the gradient coils are on, a noise of varying frequencies is heard and may be disturbing for a patient. The gradient coils are energized individually over a short time interval (a few milliseconds) to create additional fields within the static field. Because magnets attract or repel each other, the coil attempts to move in the static field, generating the “knocking” noise heard during imaging (Brant-Zawadzki 1987). The more powerful the gradient the greater the acoustical noise. The low field strength units have thus less gradient noise than the high field unit.

Computer

The MRI device is controlled by one or several computers, which set the RF- and gradient-operations and signal-colllecting necessary for the imaging sequence. They also handle the signal-and image-processing and image data display. The images are stored in a mass memory device, for example on a magnetic disk, and shown on a monitor screen. The development of MRI techniques has been dependent on the rate of development of computer power, and thus has made rapid advances in the last ten years.

The method of creating contrast between tissues

Relaxation times

The spins in the sample are excited by short RF pulses which are varied in their strength and timing to enhance the differences in the MR signal from different tissues. After the RF pulse, the
spins’ relaxation back to equilibrium are detected with a RF antenna (receiving coil) placed close to the region of interest. The signal intensity observed in an MR image is a factor of the net magnetization along the main magnetic field prior to the excitation and the rate of energy deposition from the excited spins to the surrounding lattice (relaxation time, lattice = “network” = the magnetic and thermal environment with which nuclei exchange energy in longitudinal relaxation) which is dependent on tissue type. Those tissues that allow much exchange of energy between their protons and lattice will give a more intensive signal (brighter shades of grey in image pixels).

**T1 RELAXATION TIME**

When the RF pulse is switched off, the protons realign to the main magnetic field in the z-axis. The transverse magnetization decreases (transversal relaxation); while independently, the longitudinal magnetization (z-axis) increases to its original level. The lattice absorbs energy. Longitudinal relaxation is also called ”spin-lattice relaxation”. The plot of increasing longitudinal magnetization with time is called the T1 curve, which is exponential. The rate of recovery is called the longitudinal relaxation rate and its reciprocal, T1, the longitudinal relaxation time, is the time constant of the curve. The stronger the magnetic field the longer the T1. The T1 relaxation time is shorter at lower field strengths, which allows more signal averaging with equal imaging times (Sepponen et al, 1985), (Table 1.). Field strength influences image contrast in MR imaging so that it is not possible to make direct quantitative comparisons between different T1 values at different field strengths (Rinck 1993). It is only possible in a single image to compare the intensities of the tissues with each other.

T1 is affected by the amount of water that is bound into the tissue proteins, type of protein molecules and the efficiency of the magnetization transfer between the molecule protons and water molecule protons in the hydration layer (Sepponen 1992). Pure water has a long T1 as the hydrogen nuclei have little possibility to exchange energy with other molecules. Tissues that are more proton-dense allow the exchange of energy between the protons and their lattice and have shorter T1 relaxation (= the tissue specific rate of change of magnetization). T1 of fluids and biological tissues vary from some hundred milliseconds to some seconds. It is possible to reduce T1 by administrating to the patient paramagnetic substances such as gadolinium-diethylamionetriaaminepentaacetic acid (Gd-DTPA), which is used as contrast medium in MRI.

**T2 RELAXATION TIME**

The corresponding exponential decrease in the transverse magnetic field also has a time constant, the transverse relaxation time, T2, also called ”spin-spin relaxation”. T2 is the time constant that describes the loss of the phase coherence of protons, i.e. decay of the magnetization in the transverse plane after the RF pulse. It is tissue specific. T2 is always shorter than T1 in tissues. Increase of “tissue free water” will lengthen T2. T1 is normally a few times longer than T2. Different tissues have different relaxation times because their lattices are different.

<table>
<thead>
<tr>
<th>Field strength</th>
<th>white matter</th>
<th>grey matter</th>
<th>muscle</th>
<th>fat</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.04 T</td>
<td>223</td>
<td>301</td>
<td>189</td>
<td>138</td>
</tr>
<tr>
<td>0.1 T</td>
<td>307</td>
<td>400</td>
<td>278</td>
<td>162</td>
</tr>
<tr>
<td>1.0 T</td>
<td>683</td>
<td>813</td>
<td>732</td>
<td>241</td>
</tr>
<tr>
<td>1.5 T</td>
<td>787</td>
<td>921</td>
<td>865</td>
<td>255</td>
</tr>
</tbody>
</table>

Table 1. T1 constants (in ms) at different field strengths in normal human tissue. (after Bottomley 1987)

**PULSE SEQUENCES AND TISSUE CONTRAST**

The different tissues have different T1 and T2 relaxation times and thus can be differentiated by their MR signals. The type of contrast between the tissues is controlled by the way the RF pulses are applied: their frequency, their duration, and the intervals between them. The particular combinations of these to create a desired effect that is called a pulse sequence.

The duration as well as the frequency of the RF pulse can be adjusted to tilt the magnetic vector
90° from the z-axis, or 180°. The signal given off as the magnetic vector returns to the z-axis has a constant frequency, and is called the free induction decay (FID). The differences between tissues can be amplified by sending a second RF pulse before decay is complete, thus exaggerating the differences in T1 between tissues. There are many variations of the RF pulse duration and time interval between 90° pulses, (time-to-repetition, TR). If the differences in T1 are emphasized, the pulse sequence is called T1-weighted (T1W); if it emphasizes T2 differences, the sequence is T2-weighted (T2W). T1W gives a good anatomic detail for images in a short sampling time. The contrast is low but can be enhanced by injecting Gd-DTPA as a contrast medium. The contrast agent crosses the damaged blood-brain-barrier (BBB) to show pathologic tissue by changing its signal. T2W gives high contrast for many pathological findings. The sampling time is longer than in T1W.

Noise in MR images consists of random signals that do not come from the tissues but from other sources in the machine and environment that do not contribute to the tissue differentiation. The noise of an image gives it a grainy appearance. The SNR is increased by repetition of the pulse sequences. The signal is the same on every sequence, so is added, but the noise is different, so is evenly spread and more uniform when the sequences are added. The examination time is lengthened when creating such images.

In diagnostic MR imaging, the tissue contrasts obtained are based largely on adjustment of these parameters:

- T1 contrast: The contrasts of a T1W image are based primarily on the different T1 time constants of the different tissue types.
- T2 contrast: The contrasts of a T2W image are based primarily on the different T2 time constants of the tissues.
- Proton density (PD) or intermediate contrast: The contrasts of a PDW image are based primarily on the different proton concentrations varying from tissue to tissue.

Spin echo (SE)

The most common pulse sequence is the spin-echo (SE) sequence, also called pulse echo. It is any magnetic resonance technique in which the spin echo signal is used but it is designed specially to enhance T2 differences in different tissues (Rinck 1993). An increase of water content in most pathologic changes of the brain results in lengthening of both T1 and T2 relaxation times. Prolonged T1 values decrease, and prolonged T2 values increase signal intensity in such substances as oedematous tissues, which have a high number of mobile protons. Spin-echo multissection technique, which uses a long interval between RF excitations (TR=1500-2000 ms) is useful to detect neoplastic, infectious, vascular, demyelinating, metabolic, and congenital lesions in human brain with an 0.5T MR unit (Brant-Zawadzki M et al, 1984). Modic MT et al, (1983) found SE technique with three different pulse sequence variations best for evaluation of the human spine. We tested these sequences in the dog, as those for humans could not directly be applied to the dog, because of the difference in size and possibly tissue characteristics.

The time interval between 90° pulses is called repetition time (TR = time-to-repetition). The time interval from 90° pulse to echo is called echo time (TE = time-to-echo). As TE is lengthened, T2 weighting increases, whereas lengthening of TR decreases T1 weighting without significant T2 contribution (Pauschter DM et al, 1985). Typically, long TR (over 1500 msec) and long TE (over 80 msec) give T2-weighted images, whereas short TR (under 500 msec) and short TE (under 30 msec) give T1-weighted images. When proton density (PD) images are wanted, TR is long and TE short (intermediately weighted images). (Table 2.)

In spin-echo imaging as TR controls the degree of T1 weighting and TE controls the degree of T2 weighting, varying TE and TR affect the detail and contrast of the image.

<table>
<thead>
<tr>
<th>Weighted</th>
<th>TR</th>
<th>TE</th>
</tr>
</thead>
<tbody>
<tr>
<td>T1</td>
<td>short</td>
<td>short</td>
</tr>
<tr>
<td>T2</td>
<td>long</td>
<td>long</td>
</tr>
<tr>
<td>PD</td>
<td>long</td>
<td>short</td>
</tr>
</tbody>
</table>

Table 2. T1, T2 and PD (proton density) weighting dependent on the parameters TR and TE.
T2W images may be acquired at the same time as PDW images using dual echo sequence with long TR and short TE. As the TE lengthens, the signal intensity decreases exponentially because of loss of coherence in the transverse plane. The long T2 of CSF causes minimal signal decay so that with long TE values CSF has a relatively greater signal intensity than cerebral tissue. The signal intensities in a T2 weighted image are otherwise low, but there is a high contrast between CSF and cerebral tissue (Daniels et al, 1987).

Proton density (PDW) or intermediately weighted scans are obtained by reducing the effect of longitudinal relaxation through a long repetition time (TR). Then the difference in signal after a 90° pulse is caused primarily by the difference in proton density between the tissues. When the MR signal is read out shortly after 90° pulse (short TE) a proton density weighted image is obtained (Heinrichs 1992).

In PDW images the contrast is mainly influenced by proton density (PD) of hydrogen (water, fat) in the tissue. It is weighted intermediately between T1 and T2 weighted images (long TR, short TE).

Multiecho sequence, where a multitude of echoes are excited with 180° pulses, is more efficient than single echo or inversion recovery (IR) sequence, saving examination time and creating better contrast.

To shorten imaging time, fast SE-technique (FSE = fast spin echo = RARE = Rapid Acquisition with Relaxation Enhancement) can be used. In this sequence one 90° pulse is followed by several (usually 8 to 16) 180° pulses (multiple echo spin echo sequence).

**Gradient echo (GRE)**

In the gradient echo (GRE) pulse sequence the repeated low flip angle (<90°) pulses are given at time interval TR and the echo is formed by gradient reversal operation and not by an 180° pulse, which compensates for the so called T2* effects. T2* relaxation depends on T2 relaxation time of the tissue but is affected in addition by following effects: a) magnetic field inhomogeneities caused by technical difficulties, b) local susceptibility (how easily a tissue can be magnetized) effects, c) chemical shift (the differences in resonance frequency caused by the nuclei experiencing different chemical bonds) effects. GRE sequences are sensitive to magnetic susceptibility, e.g. depicting haemorrhage and blood degradation products containing iron (Rinck 1993). Gradient echo sequence is used extensively in 0.02T, 0.04T ultralow, and 0.1T low field systems, devices mostly used in this investigation. They are called partial saturation (PS) pulse sequence in 0.1T (publication IV, Fig.1A) or saturation recovery (SR) in 0.02T and 0.04T (publ. III, Fig.1B) systems. The long gradient echo times permit the use of narrow bandwidths and enhance image contrast in low fields (0.02T, 0.04T and 0.1T). At 0.1T when the field strength is homogenous, the SNR is better with gradient echo than with spin echo sequences (Virolainen et al, 1993). Gradient echo imaging gives with a short imaging time T2W images, which look like a myelogram of the spine. The bone and disk margins are sharply delineated and with good contrast between cord and subarachnoid space.

**Types of GRE pulse sequences**

There are several gradient echo pulse sequence variants with multitude of acronyms (e.g. FAST = Fourier Acquired Steady State, FLASH = Fast Low Angle Shot, etc. imaginings). They are especially suitable for fast and ultrafast imaging because they do not wait for the 180° pulse for echo generation. In FLASH pulse sequence the RF-pulse is less than 90°. The echo is formed by inverting the magnetic field of the readout gradient.

Because of shorter TR, FLASH sequences reduce not only the scan time but also the number of slices that can be acquired. Motion artifacts thus can be reduced (Rinck 1993). Low field machines use so called FLASH-type sequences to get T2*- weighted images, when TR is long (for example TR is 2 s and TE 40-60 ms). On the other hand the same sequence can be used in 3D-imaging (for example PS3D-50/20), when T1W images are wanted.

In this investigation we used FLASH-type sequences. In the 0.02T imaging it was a variation called saturation recovery (SR) (publication I Fig.
1B, 2-4,) in 0.04T (publication I, Fig.1A; publ. III, Fig. 1, 3B) and 0.1T partial saturation (PS) (Figures in publication I-II). These differ from real FLASH-sequences because they are not necessarily used for rapid imaging but to generate T2*-weighted images with long TR (f.ex. 2s and TE~ 40-60 ms). We used the same sequence in 3D imaging to get T1-weighted images. The diagnostic advantages of 3D FLASH in intracranial tumours are based on the superior spatial resolution, as compared to conventional spinecho images. 3D FLASH gives a good T1 contrast and differentiation of grey and white matter. Gadolinium contrast can be used for further characterization of the lesion (Rinck 1995).

CBASS 3D sequence (0.1T) is based on longitudinal and transverse magnetization that are added on T2 weighting. For this a steady state effect must be created. The image will have a myelographic appearance. FIG.3.

**FIG.3. Transverse image of the thoracic spine of P43 with CBASS 3D sequence TR 16/TE 8, FA 90°, FOV 256x256, matrix 256x256, slice thickness 3 mm, imaging time 7 min 45 sec. The bright signal of CSF and epidural fat around the spinal cord (arrow) give a myelographic appearance.**

Inversion recovery (IR)

The inversion recovery sequence is designed to enhance differences in T1 between different tissues, for example between grey and white matter in brain ( Doyle et al, 1981). In IR pulse sequence a 180° pulse is given first, and after a time interval T1 (time-of-inversion), a 90° and thereafter a 180° pulse as in SE-sequence for echo formation. The echo can also be formed with the gradient reversal technique. IR three dimensional (3D) technique provides high T1 contrast in low field strength imaging (publ. IV, Fig.1C). Low field strength imaging with gradient echo technique with IR sequence has the advantage that with GRE a better SNR is gained than with SE, because a narrow signal sampling band or long echo sampling time can be used.

**Saturation recovery (SR)**

Saturation means that equal numbers of spins are aligned against and with the magnetic field, so that there is no net magnetization (equilibrium state). Signals with different relaxation characteristics can be measured by two sequences of RF pulses for each projection, resulting in different types of images. A so-called saturation recovery (SR) image is reconstructed from echoes found in the first sequence, whereas the so-called inversion recovery image contains data from the second sequence (Go et al, 1983).

A type of partial saturation pulse sequence is a sequence in which the preceding pulse leaves the spins in a state of saturation so that the recovery has taken place at the time of the next pulse.

Saturation recovery images exhibit poor contrast between grey and white matter and high signal intensity from fatty tissues (Go et al, 1983).

**FACTORS AFFECTING THE IMAGE**

**Image contrast in MRI**

In MRI there is no absolute signal reference and the diagnostic evaluation of an image is based on the observation of the differences in intensities within a single image. A large number of RF pulse sequences have been developed in human MRI to exploit different contrast mechanisms. None of the sequences or proposed protocols has demonstrated a general superiority over the others (Baleriaux et al, 1995).

Contrast in MR images is mainly a result of the differences in proton density and the relaxation times T1 and T2. The contrast between two different tissues can be optimized with variations of the following parameters: TR, TE, time-to-inversion, flip angle (FA). For example partial saturation (PS) pulse sequence can generate
PDW and T1W images with increased contrast between regions with different relaxation times.

Blood flow, diffusion, and perfusion are other factors influencing the signal emitted by the structures within a volume element or voxel. The voxel and pixel sizes influence spatial resolution and contrast. A large voxel can contain signals from different tissue structures and their average determines the intensity of the corresponding pixel seen on the monitor (partial volume effect). The smaller the voxel can be kept, the fewer different structures are represented in a single pixel, and the better the spatial resolution that is achieved (Rinck 1993).

**TISSUE DIFFERENTIATION**

In the spine, on a T1W image, proton-poor cortical bone can be seen to frame vertebral bodies with higher signal homogenous marrow present centrally. T1 images will show high signal (white) from fat, subacute hemorrhage, and protein rich fluid. Dark signal is present in T1 image from bone, calcium, CSF and cartilage. Other tissues tend to be of intermediate or isointensive nature.

**Signals from flowing fluid**

Vessels may be dark, relating to a lack of signal caused by rapid outflow of blood from the slice between the 90° and 180° pulses (diminished saturation), or be light, due to increased signal intensity with the inflow of blood to the slice that has not been previously excited between 90° pulses. This seems paradoxical but it is simply a factor of velocity of flow and saturation time. At higher velocities the intensity of the flowing blood in the vessels drops because the nuclei have not sufficient time to get saturated because of their speed and turbulence. At low flow there is an increase in intensity caused by the influx of nuclei into the imaged volume (Crooks et al, 1980). Slow flowing CSF appears dark in T1W images and bright in T2 weighted images. (FIG 10.A,B.) Relatively fast flowing CSF behaves like flowing blood in PD, T1 and T2W images.

**GADOLINIUM (Gd-DTPA) AS A PARAMAGNETIC CONTRAST AGENT IN MRI**

The diagnostic sensitivity of MRI can be improved with paramagnetic substances as contrast media. Ions and molecules that contain unpaired electrons demonstrate a paramagnetic behaviour when placed in an external magnetic field. They hasten relaxation rates of protons in their microchemical environment (Brasch, 1983; Mendonca-Dias et al, 1983; Brasch et al, 1984; Weinmann et al, 1984). Small amounts of paramagnetic substance shorten markedly the T1 relaxation time, so a low dose gives the desired enhancement through shortening T1 time of the tissue and therefore increasing signal intensity. T2W images are influenced only with high dose. A shortened T2 would give a decreased signal intensity (Paajanen et al, 1986), and this is of no value.

Gadolinium ions (Gd³⁺) from the lanthanide series of rare-earth elements contain seven unpaired electrons. These electrons provide gadolinium with strong paramagnetic properties (Pople et al, 1959). Gd³⁺ can be chelated with diethylaminitraminepentaaetetic acid (DTPA) to form a stable complex (Gd-DTPA). After intravenous application it distributes primarily in the intravascular, extracellular space and quickly passes into the extravascular, interstitial space (Brasch et al, 1984). The leakage creates a contrast enhancement that is seen in T1W images by shortening the T1 value.

Gd-DTPA enhances the border between certain brain tumours and normal tissue (Carr et al, 1984; Felix et al, 1985). It gives also a better distinction between oedema and brain tumour and helps in the diagnosis of brain abscesses and infarcts (Brasch et al, 1983; Runge et al, 1985a). The gadolinium is not visualized directly but indirectly by the influence of paramagnetic gadolinium on the protons of the tissue (Gavin 1994).

**IMAGE GENERATION**

The three planes of the body are located by applying a magnetic gradient in each plane and sending a RF signal. Where the three planes
intersect for the same signal, a voxel is created with its own location and emitted signal.

In order to select a slice from the object one must switch on one of the gradients (z) and transmit to the object an RF pulse which contains a well-defined band of frequencies. This results in slice selection because only those spins, which resonate with the RF-pulse are affected (Pople 1959). With additional gradient operations in y and x directions it is possible to encode the spatial information into the phase and frequency of the resulting NMR signal from the selected slice.

This encoded spatial information is extracted from the NMR signal by means of Fourier transformation, i.e., by performing a frequency analysis on the signal data. This transforms the complex waveforms into a single number, their frequency, and it is this number which is used, thus speeding up the processing to a realistic time.

This procedure results in a three dimensional matrix of voxels, in which the intensity of each voxel reflects the magnetic properties and proton density of the tissue in that location. The actual image is displayed as a two-dimensional pixel matrix one voxel deep. The pixel determines the inplane resolution of the image, the voxel size takes into account the slice thickness too. The smaller the voxel, the higher the resolution, but a small voxel gives a weaker signal and thus a lower SNR.

Because their signal coding is performed in all three dimensions, modern MRI scanners enable the free selection of slice orientation f. ex. sagittally, transversally (axially), or dorsally (coronally in human imaging) after the scan.

Slice thickness and interslice gap

The slice thickness is determined by the frequency bandwidth (BW) of the excitation pulse and the strength of the slice-defining gradient (Gz). The section thickness is decreased by either narrowing the BW or increasing Gz (Kelly 1987).

The excitation of adjacent slices provokes cross-excitation between them. This reduces the time available for remagnetization of tissue in adjacent slice locations and results in decreased SNR and adds T1 contrast on T2 images. Therefore some older MR imagers have an obligatory interslice gap, which may decrease lesion detectability (Pauschter et al, 1985). An interslice gap of about 50% of the slice thickness is a compromise. But for example a gap of 2.5 mm between 5-mm slices can lead to missing a lesion smaller than 2.5 mm. This problem existed when the ultralow field strength 0.02T ("Acutscan™") and 0.04T ("Magnaview™") imagers were in use. The more modern scanners have eliminated this problem.

Partial volume averaging

The pixel is the 2D representation on the image of a voxel, the volume element. If the tissue is homogenous the signal intensity (shade of grey) of a pixel and its corresponding voxel represent the tissue accurately. But in inhomogeneous samples the signal intensity of a pixel is the average of the different MR properties of the corresponding voxel, not necessarily typical of any of its components. This phenomenon is called partial volume averaging (PVA). The PVA may lead to false interpretations, for example in the comparison of normally bilaterally symmetrical regions an asymmetric signal intensity area is discovered and diagnosed as a lesion when a patient’s head is obliquely angled (Kelly 1987). An example of partial voxel averaging is demonstrated in the FIGURES 12, D and E.

The smaller the voxel, the better the spatial resolution and contrast (Rinck 1993). On the other hand the bigger the voxel size the stronger the signal gained from each voxel. Smaller lesions may be missed using small voxel size due to weak signal despite the good resolution. MRI with surface coil allows thin sections with high spatial resolution, which reduces partial volume averaging (Pauschter et al, 1985). This was a reason to develop a surface coil especially for dogs.

ARTIFACTS

Artifacts may mimic pathological changes or make the image unreadable. They have been categorized into four main groups by Johnson et al, (1989):

1: magnetic field perturbations
2: RF artifacts and gradient related artifacts
3: motion and flow artifacts
4: signal processing and mapping artifacts.

Motion artifacts may be a problem when imaging dogs. Even if the head or spine does not move with breathing the sedation may not be deep enough to control tremor or its duration may be too short for the imaging time which can last up to one hour with the ultralow field imagers. Only anesthesia machines that have no ferromagnetic parts can be used near the scanners and none was available. We therefore needed to use a simple and reliable form of sedation.

SAFETY OF MRI

Hazards in magnetic resonance imaging can arise because of static magnetic fields, varying magnetic fields, radiofrequency fields and cryogens.

The static magnetic field can create acute hazards in the form of ferromagnetic objects behaving like projectiles in the range of magnetic field. This danger is greater at higher field strengths. A high field system needs a heavy shielding of walls. Ultralow- to mid field strength systems have a limited stray field and the shielding can be correspondingly light. This has the advantage of lower building costs. Metal surgical implants and clips can become dangerous in the magnetic field as they may move or the large ones may be heated. Radiofrequency fields of the MR-imager disturb the function of pacemakers in the examination room. The rules for handling and management of cryogen in high field units must be strictly followed to avoid accidents.

Subacute risks may arise from exposure to magnetic radio-frequency fields. Until now no damage has been found when units up to 2T have been used. However it can be recommended that unnecessary exposure of the examiners should be avoided. The noise level created by switching of the gradients increases with the field strength and can be very loud and unpleasant. The persons watching over the animal patients in a high field examination room should use hearing protection. Low field units have low noise level. Even these noise changes when switching on different gradients may scare inadequately sedated dogs.

The risks are greater in connection with high field systems but all persons using or entering any MRI examination room should be informed about potential risks even when no special danger is proven to exist (Rinck 1990). The Finnish Center for Radiation and Nuclear Safety supervises the use of MRI in Finland (Huurto et al, 1993).
AIMS OF THE PRESENT STUDY

The main purpose of the present work was to study the application of MRI in canine neurological disorders. The specific aims of the study were:

1. To compare the suitability of the use of ultralow-, low field strength, and high field strength magnetic resonance imagers by studying spontaneous CNS diseases in dogs.

2. To find out the practical solutions for MRI procedure of canine patients concerning brain and spine MRI: immobilization, positioning, and local coil application.

3. To improve the resolution and diagnostic effectiveness of MRI for canine patients by evaluating imaging parameters, and contrast application.

4. To improve coil design for better canine brain imaging resolution.

5. To evaluate the usefulness of MRI for veterinary medicine.
MATERIALS AND METHODS

MATERIALS

Normal dogs

Ten dogs (N1-10) were clinically healthy normal dogs. These dogs had no neurologic abnormalities at the neurological examination. No laboratory or radiographic studies were made in these dogs.

Three normal dogs were imaged with three different coils in the evaluation of the new brain coil designed by us and described in publication V. Seven clinically normal dogs (N1-4, 6-8) underwent 8 MRI studies as control dogs for brain imaging (Table 3). The mean age for the normal dogs was 7 years (range 2 to 10 years) and the average weight was 13 kg. One clinically normal English setter (N4) underwent an 0.04T and one Pointer (N2) underwent an ultralow field imaging with 0.02T scanner. Another Pointer (N3) was examined with low field strength (0.1T) and high field strength (1.0T) scanners and one German Pointer (N1) with only the high field strength scanner.

One normal German pointer (N1) served as control dog for cervical spine imaging and one Rottweiler for thoracic spine imaging. Five normal dogs (N2, N3, N6, N7, N8) two Pointers, one Rottweiler, one Finnish hound, and one German shepherd were control dogs for lumbar spine imaging.

The dogs, scanners and contrast studies used are listed in Table 3.

Patients

The decision for imaging dog patients with signs of CNS disturbances was made together with the owner, referring veterinarian and the author. Fifty-six (56) patient dogs were examined with MRI. They were patients of the Department of Clinical Sciences, College of Veterinary Medicine (later University Animal Hospital, Faculty of Veterinary Medicine, University of Helsinki) for different neurological disorders. Their age range was from 2.5 months to 11 years. The weights varied from 2.75 kg.

Four patients had both low-and high field-strength examinations and one patient had twice a low field strength scan. So all together 72 MR studies of patients were done on 56 dogs. There were 34 patients with brain disease, 9 patients with cervical spine disease, 3 patients with thoracic spine disease, and 11 patients with lumbar spine disease.

Patients with suspected brain lesions

Thirty-four patient dogs (P1-34) with various neurological signs underwent 36 MRI examinations of the brain. Twentyone breeds and one mongrel were represented. Fifteen were males, 18 females and one a neutered female. Ages of the dogs were from 2.5 months to 10 years. Weight range varied from 2 to 75 kg. The patients came to the clinic because of different neurological disorders indicating brain disease.

Patients with suspected spinal lesions

Twenty-three patients (P22, P35-56) with neurological signs related to diseases in the spine underwent 23 MRI examinations of these areas. Ten different breeds were represented. Eight were females (one of them neutered) and 15 males. Their age range was from 6 months to 11 years. Their weight varied from 9 to 48 kg.

Cervical spine. The MRI examination of the cervical spine was done in 9 dog patients 10 times as one dog was examined both with low- and high field strengths. One normal dog’s cervical area was scanned as a control.

Thoracic spine. The thoracic spine was imaged in one normal dog and three patients.

Lumbar spine. Five normal dogs of four breeds (N2-3, 6-8), and 11 patients (P46-56) of seven different breeds underwent 16 MRI studies. The age range of normal dogs was from 4 to 9 years. Weight range was from 16 to 45 kg. The age range of the patients was 4 to 10 years. Their weight varied from 9 kg to 45 kg. The examination of the lumbar spine was done in 11 patient dogs eleven times. Five normal dogs were imaged for
### Table 3. The field-strengths, areas, and contrast of MRI used in this investigation.

<table>
<thead>
<tr>
<th>Breed</th>
<th>Nr.</th>
<th>Sex</th>
<th>Weight kg</th>
<th>Age at MRI</th>
<th>MRI</th>
<th>Magnevist®</th>
<th>Area</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal dog</td>
<td>N1</td>
<td>m</td>
<td>28</td>
<td>10 years</td>
<td>1.0T</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Normal dog</td>
<td>N2</td>
<td>f</td>
<td>16</td>
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<td>0.02T</td>
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<td></td>
</tr>
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<td>Normal dog</td>
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<td>m</td>
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<td></td>
</tr>
<tr>
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<td>m</td>
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<td>2 *</td>
<td>0.04T</td>
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</tr>
<tr>
<td>Normal dog</td>
<td>N5</td>
<td>f</td>
<td>11</td>
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<td></td>
<td></td>
</tr>
<tr>
<td>Normal dog</td>
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<td>14</td>
<td>5 *</td>
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<td></td>
<td></td>
</tr>
<tr>
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<td>N10</td>
<td>m</td>
<td>13</td>
<td>3 *</td>
<td>0.1T</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Patients</td>
<td>P1</td>
<td>f</td>
<td>28</td>
<td>10 *</td>
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<td>0.2 ml/kg</td>
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</tr>
<tr>
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<tr>
<td>Griffon Bruxellois</td>
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<td>P8</td>
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<tr>
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<td>P9</td>
<td>m</td>
<td>17</td>
<td>10 *</td>
<td>0.04T</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Collie, lm</td>
<td>P10</td>
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<td>20</td>
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<tr>
<td>Collie, lm</td>
<td>P11</td>
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<td>20</td>
<td>10 *</td>
<td>0.04T</td>
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<tr>
<td>German pointer</td>
<td>P12</td>
<td>f</td>
<td>28</td>
<td>8 &amp; 8 years</td>
<td>0.04T + 1.0T</td>
<td>brain, neck</td>
<td></td>
</tr>
<tr>
<td>Golden retriever</td>
<td>P13</td>
<td>m</td>
<td>38</td>
<td>6 *</td>
<td>0.04T</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Golden retriever</td>
<td>P14</td>
<td>m</td>
<td>30</td>
<td>2 *</td>
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<td></td>
</tr>
<tr>
<td>Beagle</td>
<td>P15</td>
<td>f</td>
<td>15</td>
<td>10 *</td>
<td>0.04T</td>
<td></td>
<td></td>
</tr>
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<td>Miniature schnauzer</td>
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<td>f</td>
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<td>6 years</td>
<td>1.0T</td>
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</tr>
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<td>Dachshound</td>
<td>P17</td>
<td>f</td>
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<td>6 *</td>
<td>0.04T</td>
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<td></td>
</tr>
<tr>
<td>German shepherd</td>
<td>P18</td>
<td>m</td>
<td>36</td>
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<td>brain, brain</td>
</tr>
<tr>
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<td>P19</td>
<td>f</td>
<td>38</td>
<td>2 *</td>
<td>0.04T</td>
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<td></td>
</tr>
<tr>
<td>Japanese chin</td>
<td>P20</td>
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<td>3</td>
<td>1 year</td>
<td>0.04T</td>
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<td></td>
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<tr>
<td>Rottweiler</td>
<td>P21</td>
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<td></td>
</tr>
<tr>
<td>Rottweiler</td>
<td>P22</td>
<td>m</td>
<td>40</td>
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<td>0.2 ml/kg</td>
<td></td>
</tr>
<tr>
<td>Rottweiler</td>
<td>P23</td>
<td>f</td>
<td>38</td>
<td>2 yrs</td>
<td>1.0 T</td>
<td></td>
<td></td>
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<tr>
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<td>P24</td>
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<tr>
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<td>P25</td>
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<td>10</td>
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<tr>
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<td>11</td>
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<tr>
<td>Newfoundland dog, lm</td>
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<tr>
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<td>m</td>
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<tr>
<td>Newfoundland dog, lm</td>
<td>P29</td>
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<td></td>
</tr>
<tr>
<td>Poodle</td>
<td>P30</td>
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<td>7</td>
<td>4 yrs</td>
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</tr>
<tr>
<td>German shepherd</td>
<td>P31</td>
<td>m</td>
<td>35</td>
<td>7 yrs</td>
<td>0.1T + 1.0T</td>
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<td>brain, brain</td>
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<tr>
<td>St. Bernard dog</td>
<td>P32</td>
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<td>0.2ml/kg</td>
<td></td>
</tr>
<tr>
<td>Finnish hound</td>
<td>P33</td>
<td>f</td>
<td>15</td>
<td>6 mo</td>
<td>1.0T</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Mongrel</td>
<td>P34</td>
<td>f/n</td>
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Table 3 cont. The field-strengths, areas, and contrast of MRI used in this investigation.

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</tr>
<tr>
<td>Cervical spine</td>
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<tr>
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<td>brain, neck</td>
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<td>18</td>
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<td>33</td>
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<tr>
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<td>42</td>
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<tr>
<td>P38</td>
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<td>39</td>
<td>5 &quot;</td>
<td>1.0T</td>
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<tr>
<td>P39</td>
<td>Rottweiler</td>
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<tr>
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<tr>
<td>P22</td>
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<td>m</td>
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<td>0.1T + 0.1T</td>
<td>brain, neck</td>
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<tr>
<td>Thoracic spine</td>
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<tr>
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<td>1.0T</td>
<td>thor., lumber</td>
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<tr>
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<td>Lumbar spine</td>
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<tr>
<td>Normal dog N6</td>
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<td>2 &quot;</td>
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<td>9 &quot;</td>
<td>0.02T</td>
<td></td>
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<td>Normal dog N8</td>
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<td>45</td>
<td>4 &quot;</td>
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<td>45</td>
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<td>P56</td>
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<td>10 &quot;</td>
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WH = wirehaired, f = female, f/n = female, neutered, m = male, T = Tesla, lm = littermate

Table 4. The number of MRI studies with different field strengths.

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<th>MRI object</th>
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<td></td>
<td>ultralow</td>
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<td>44</td>
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<td>4</td>
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<tr>
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<td>2</td>
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<td>1</td>
<td>17</td>
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<tr>
<td>Total</td>
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<td>33</td>
<td>20</td>
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Table 5. Summary of the sequences used at ultralow field strength imagings of brain.

<table>
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<tr>
<th></th>
<th>40 dogs</th>
<th>Range of TR (time-to-repetition)</th>
<th>Range of TE (time-to-echo)</th>
<th>Range of slice thickness in mm</th>
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<tbody>
<tr>
<td>T1W</td>
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<td>75-500</td>
<td>25-60</td>
<td>5-10mm</td>
</tr>
<tr>
<td>T2W</td>
<td></td>
<td>1000-2600</td>
<td>60-130</td>
<td>7-10mm</td>
</tr>
<tr>
<td>PDW</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

T1W = T2 weighted, T2W = T2 weighted, PDW = proton density weighted

Table 6. Summary of the sequences used at low field strength imagings of brain.

<table>
<thead>
<tr>
<th></th>
<th>14 dogs</th>
<th>Range of TR (time-to-repetition)</th>
<th>Range of TE (time-to-echo)</th>
<th>Range of slice thickness in mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>T1W</td>
<td></td>
<td>50-125</td>
<td>14-40</td>
<td>5-9mm</td>
</tr>
<tr>
<td>T2W</td>
<td></td>
<td>1000-2700</td>
<td>60-250</td>
<td>2.5-7mm</td>
</tr>
<tr>
<td>CBASS</td>
<td>15-16</td>
<td>5/7/08</td>
<td>30-120</td>
<td>2-3mm</td>
</tr>
<tr>
<td>PDW</td>
<td>2500-4000</td>
<td></td>
<td></td>
<td>5-6mm</td>
</tr>
</tbody>
</table>

T1W = T2 weighted, T2W = T2 weighted, PDW = proton density weighted, CBASS = completely balanced steady state sequence
### Table 7. Summary of the sequences used at high field strength imagings of brain.

<table>
<thead>
<tr>
<th>Weighting</th>
<th>14 dogs</th>
<th>Range of TR (time-to-repetition)</th>
<th>Range of TE (time-to-echo)</th>
<th>Range of slice thickness in mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>T1W</td>
<td></td>
<td>55-2500</td>
<td>15-25</td>
<td>4-5 mm</td>
</tr>
<tr>
<td>T2W</td>
<td></td>
<td>330-2500</td>
<td>15-90</td>
<td>2.5-7 mm</td>
</tr>
<tr>
<td>PDW</td>
<td></td>
<td>2500</td>
<td>23-25</td>
<td>5-6 mm</td>
</tr>
</tbody>
</table>

T1W = T2 weighted, T2W = T2 weighted, PDW = proton density weighted

### Table 8. Summary of the sequences used at ultra low field strength imagings of spine.

<table>
<thead>
<tr>
<th>Weighting</th>
<th>14 dogs</th>
<th>Range of TR (time-to-repetition)</th>
<th>Range of TE (time-to-echo)</th>
<th>Range of slice thickness in mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>T1W</td>
<td></td>
<td>100-400</td>
<td>40</td>
<td>5 - 9 mm</td>
</tr>
<tr>
<td>T2W</td>
<td></td>
<td>1000-1500</td>
<td>60-80</td>
<td>9 mm</td>
</tr>
<tr>
<td>PDW</td>
<td></td>
<td>2500</td>
<td>15</td>
<td>4 mm</td>
</tr>
</tbody>
</table>

T1W = T2 weighted, T2W = T2 weighted, PDW = proton density weighted
### Table 9. Summary of the sequences used at low field strength imagings of spine.

<table>
<thead>
<tr>
<th>Weighting</th>
<th>14 dogs</th>
<th>Range of TR (time-to-repetition)</th>
<th>Range of TE (time-to-echo)</th>
<th>Range of slice thickness in mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>T1W</td>
<td></td>
<td>155-500</td>
<td>15-20</td>
<td>5 mm</td>
</tr>
<tr>
<td>PS 3D</td>
<td></td>
<td>60-155</td>
<td>20-25</td>
<td>3 mm</td>
</tr>
<tr>
<td>T2W</td>
<td></td>
<td>2500</td>
<td>90</td>
<td>3-5 mm</td>
</tr>
<tr>
<td>CBASS 3D</td>
<td></td>
<td>16</td>
<td>8</td>
<td>3 mm</td>
</tr>
<tr>
<td>PDW</td>
<td></td>
<td>2500</td>
<td>15</td>
<td>4 mm</td>
</tr>
<tr>
<td>IR</td>
<td></td>
<td>500(T1 350)-800(T1 90)</td>
<td>26</td>
<td></td>
</tr>
</tbody>
</table>

T1W = T2 weighted, PS = partial saturation, IR = inversion recovery.

### Table 10. Summary of the sequences used at high field strength imagings of spine.

<table>
<thead>
<tr>
<th>Weighting</th>
<th>14 dogs</th>
<th>Range of TR (time-to-repetition)</th>
<th>Range of TE (time-to-echo)</th>
<th>Range of slice thickness in mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>T1W</td>
<td></td>
<td>300-500</td>
<td>18-20</td>
<td>3-5 mm</td>
</tr>
<tr>
<td>T2W</td>
<td></td>
<td>2500</td>
<td>90</td>
<td>3-5 mm</td>
</tr>
<tr>
<td>PDW</td>
<td></td>
<td>2200</td>
<td>80</td>
<td>3 mm</td>
</tr>
</tbody>
</table>

T1W = T2 weighted, PDW = proton density weighted.
comparison in lumbar spine. The number of studies and the different field strengths are summarized in Table 4.

**MR imaging devices**

**Ultrasound field strength**

0.02T: “Acetscan™”, (Instrumentarium Imaging Corp, Finland)

These ultralow field strength imagers had resistive magnets and were water chilled. The size of the aperture was 75 cm. The resonance frequency was 833 kH. The slice thickness was 10 mm and the image matrix 256x128. There were six preprogrammed sequences and five programmable user sequences. In these latter sequences the user could change various parameters. The field-of-view of the standard head coil was 30 cm allowing spatial resolution of 3 to 5 mm, examination time was 60 - 80 min. The patient couch had a light projector with an electronic position indicator as also other field strength machines. 0.02T was the lowest magnetic field strength that has been used in clinical MRI for human patients (Korman et al, 1987).

0.04T: “Magnaview™”, (Instrumentarium Imaging Corp, Finland)

The other ultralow field imager had a resistive magnet, the type was solenoid with end correction coils (shim coils). The bore was 71 cm, the weight was 850 kg.

**Low field strength**

0.1T: “Mega 4HT™”, renamed “Merit™”, from Instrumentarium Imaging Corp, later Picker Nordstar Inc.

They had a resistive magnet and closed loop water-cooling. The diameter of the inner bore was 65 cm.

“MeritTM” has the following standard pulse sequences:

1. Spin echo techniques were combined with bandwidth optimization, based on the relaxation properties of tissues. The result is improved contrast and SNR.

2. Fast spin echo techniques provided the high image quality of conventional spin echo in reduced scan times.

3. Gradient echo techniques at low field could be used in completely new areas due to the high absolute homogeneity of the magnet. The long gradient echo times permitted the use of narrow bandwidths and enhance image contrast.

4. IR and STIR techniques, recognized for their sensitivity, were available for routine use with Turbo Multiple Slice technique. IR3D technique provided high T1 contrast and contiguous slices.

5. Steady State techniques provided excellent contrast-to-noise ratio in minimal acquisition time.

**High field strength 1.0T scanner**

The high field strength scanners had a superconducting magnet with 1.0T main magnetic field. The size of the inner diameter of the bore was 60 cm in “Magnetom 42SP/42SP™” and 90 cm in “Magnetom Impact™”. They had a large variety of preprogrammed sequences for human imagings.

**METHODS**

**Neurological examination**

For all the patients a neurological examination was done according to the protocol “Neurological examination” of the Veterinary College in Helsinki. This includes posture and movement observation, and tests of cranial and spinal nerve function.
Radiographic examination

A radiographic examination of the head was made in ten dogs with neurological signs with plain films in lateral and cranio-caudal projections and in some cases supplemented with an open mouth rostro-caudal view for evaluation of inner ear and bulla tympanica regions, and in oral-aboral position for evaluation of foramen occipitale. In nine cases CSF was taken from the cisterna magna for examination. Myelography by cisterna magna or intervertebral spaces of L5-L6 or L6-L7 was done with iohexol\(^1\) (0.25-0.5 ml/kg).

Immobilization

An anesthesia machine could not be used use because we had no specially designed nonferromagnetic anesthesia machine of our own. It was therefore necessary to try sedative drugs to give complete immobilization for the required scanning time.

The dogs were canulated and sedated for all MRI examinations. Medetomidine\(^2\) alone or combined with methadone\(^3\) was used. For sedation 40 to 60 \(\mu\)g/kg medetomidine was given iv or im. This usually was sufficient for about thirty minutes of immobilization. If more scanning time was needed half of the initial dose of medetomidine was injected intramuscularly before a new imaging sequence began. In 49 dogs medetomidine was combined with methadone 0.5-1 mg/kg. The dosage of medetomidine was the same as used alone and equal to that of methadone in milliliters. In 18 dogs medetomidine was combined with ketamine\(^4\) (4-6 mg/kg). Atipamezole\(^5\), an antidote for medetomidine, was given im if needed after the imaging. The dose for atipamezole was equal in milliliters to the given amount of medetomidine.

Positioning for the examination

General

Each patient was positioned individually, to adjust for the anatomy and the type of the local coil. Careful positioning was necessary for optimal imaging results. The head or spinal region was placed as near as possible and in center of the closest fitting local coil. There were some specific positioning required for each machine:

- Ultralow field units: A standard human-type spinal surface coil was the only available for this imaging. It was placed on the patient couch under the dog. The dogs were monitored by directly staying with them or observing them through a glass window.
- Low field units: The standard flexible multipurpose coils and the dedicated canine brain coil (publication V, Fig 2) could be wrapped around the spine or head. The heads of the dogs were imaged in lateral or sorial recumbency with standard head or knee coils, FIG 2.A.D. The standard multipurpose and brain coils were positioned around the region of interest. For spinal imaging with standard spine coils designed for human patients the dogs were in lateral recumbency to minimize the artifacts caused by breathing movements and for closer contact with the coil placed under them.
- In the room for this machine we monitored the patients directly in their immediate vicinity or with a video camera located in the examination room.
- High field units: The coils we used were human head or knee coils for brain and standard spinal coil for spinal imaging. These were used in the same manner as with low fields. No flexible multipurpose coils were available. The dogs were monitored with a videocamera as with the low field unit.

Spine

**CERVICAL SPINE**

The dogs were positioned in lateral recumbency on the patient couch. The neck area was extended and supported with foam plastic cushions. A loop shaped flexible surface coil was fitted tightly on the lateral side of the neck of the

---

\(^1\)Omnipaque® 300 mg/ml, Nycomed Imaging AS, Oslo, Norway.

\(^2\)Domitor®, Orion-Farmos, Turku, Finlannd.

\(^3\)I-Polamivin®®, Schering Lrd.

\(^4\)Ketalar® 10 mg/ml®, Parke-Davis.

\(^5\)Antisedan “®, Orion-Farmos, Turku, Finland.
patient in the low field imaging. A surface coil was centered on the area of suspected lesion. One patient with signs of cervical spinal compression was placed for the low field strength examination in left lateral recumbency, in that position with the head flexed towards the chest for provocation of cervical vertebral subluxation and scanned.

**Lumbar Spine**

The dogs were positioned in all cases but one in dorsal and lateral recumbency on top of a human spinal coil. One control dog was positioned in sternal recumbency for the low field imaging and a loop shaped flexible local coil was fitted on the dog's lumbar area (FIG.4). This was a try to improve the scan of lumbosacral spine with this method that was not available earlier. Sagittal T1 and T2 weighted images were taken of all dogs. In addition transverse T1 slices were made of two control dogs (N3, N7) using 0.1T and 1.0T correct positioning and determine the image slices to be made. For images of the spinal cord a dorsal scout image was done to determine the sagittal slices to be made. FIG.5.A,B,C.

**FIG 4.** A loop shaped flexible multipurpose coil for lumbar spine imaging.

Imaging planes

The imaging planes were called sagittal, transversal (called also axial), and dorsal according to the recommendation of the International Veterinary Radiology Association/MRI section. Additionally, modifications of these were used if a lesion was projected better that way for example, the transverse-oblique scan that was used in two dogs in poor condition to save time. The requirement for at least two imaging planes for demonstration of anatomy or pathologic changes was met in all but these two dogs. In brain MRI a sagittal localizer series, a so-called “scout image” was done first to ensure the slices to be made.

**MRI sequences in brain examination**

**Ultralow field strength**

At each examination, up to five different sequences depending on the time available and the condition of the patient were used and evaluated for image quality. After experience was gained, fewer protocols were tested on each machine. They are described in Publications I-III. Table 5. shows the range of sequences used in imagerings.
Low field strength

As the same physicist who had tested the sequences in ultralow field machines made also these scans his experience helped to find quickly the suitable sequences for low field strength. They had to be designed to show the clinically suspected lesions. Usually if the images of the first sequence were not of good diagnostic quality new sequences were tested. They are described in Publications IV-V. Table 6. shows the range of sequences used in imagings.

High field strength

In high field strength imaging in the contract facility of a private company (Helsingin Lääkärikeskus, Teslamed, Höylämötie 14, Helsinki) the local specially trained radiographers operated the machines. The preprogrammed sequences could not be used. They had to create a suitable sequence together with the author who had an idea what to expect to see in each dog. After the first sequence new sequences were tried out one by one if needed for diagnosis. Table 7. shows the range of sequences used.

Contrast medium

In 6 patients low- and high field strength T1 weighted transverse images were taken immediately after intravenous injection of gadolinium-DTPA 0.1-0.2 mmol/kg. (Table 3.) Eleven patients were given gadolinium-GDTPA 2) (‘Magnevist”, Schering Ltd., Germany) (0.1-0.2 mmol/kg) through an intravenous catheter in v. cephalica as an intravenous bolus immediately before repeating of the T1 sequences of brain or spine. The dogs were cannulated before sedation. The aim was to see a possible BBB leakage to confirm a lesion. No contrast studies were done in patients in very poor condition requiring a scanning time as short as possible.

The pre-contrast T1 weighted images were compared with the post-contrast T1 weighted images.

Recording of the images

The images were recorded on films for diagnostic evaluation.
Results

THE EFFECTIVENESS OF IMMOBILIZATION

The immobilization with sedation as described in the Methods was adequate to immobilize dogs even during the longest imaging times. The sedation could be lengthened with repeated injections of sedatives if the dog showed signs of awakening during the examination. Close observation of the dogs was easier in the ultralow- and low field strength units because of the larger and shorter bore, clearer (more light in the bore) patient monitor picture, and the shorter distance to the patient than in the high field unit. Staying in the imaging area of a low field unit was also more comfortable for the observing veterinarian because of the lower acoustic noise level compared with the high field unit. The noise of any unit did not disturb the patients. The possibility to use an antidote for quick recovery from the sedation for immediate transportation was safe for the dogs and practical for the owners as they could take them home after imaging.

THE POSITIONING OF THE DOGS

The positioning was initially based on the standards for humans, but was soon modified because of the different shape of the head and spine of the dog.

Positioning for the brain imaging

Positioning in sternal recumbency provided the best symmetry for brain imaging. Lateral recumbency made it very difficult to fix the head in an exactly straight position because the loose skin of the dogs hindered accurate positioning. This was particularly a problem when using human knee coils.

Positioning for the spinal imaging

The patient’s position could not easily be adjusted to straighten the spine. Dorsal recumbency without intubation may make a dog’s breathing difficult and so this positioning was not used because of this risk. The dogs were not intubated because that would have required deeper sedation to the level of light anaesthesia. The dogs were supported with foam cushions to get the spine as straight as possible for an exact mid-sagittal view. The cervical, thoracic and lumbar spine had each to be extended for different scanings of sagittal images because of the curvature. Because of the anatomical curving of the spine from the straight midline only short sections of the spine could be seen on one image. The cranial and caudal cervical vertebrae, the cranial thoracic, the caudal thoracic area and the lumbar spine required each a separate sagittal image. Exact localization of a lesion in the sagittal plane was needed before transverse imaging was begun so the necessary transverse slices could be limited to the affected region. Otherwise the required time for transverse imaging would have been too long if large parts of the spine had to be examined with numerous slices. The localization of a disk in transverse scans had to be very exact for correct angling of the slice through the intervertebral space. The dorsal slices were useful as orientation scans for sagittal imaging and occasionally for more complete anatomic localization for possible surgery (see FIG.22.B.).

THE RESULTS OF BRAIN MRI

T1W images of normal dogs

T1W imaging with different field strength scanners revealed normal cerebral, cerebellar and pontine macroscopic anatomy in the sagittal plane (see FIG.6.C.)

Brain tissue had intermediate (grey) signal intensity. The white matter was slightly lighter than the grey matter. CSF, cortical bone, and air had very low signal intensity (black) in T1W images created with all the used imagers.

The fat in bone marrow of the cranial bones and orbit gave high intensity signals appearing bright white. The air in the frontal sinuses and the nasal
cavity was black because of signal void. The muscles of the head and neck consisted of tones of grey of moderate signal intensity. The anatomy of the orbit was clearly visible with all field strengths.

In the transverse plane at the level of the cerebrum the lateral ventricles, the third ventricle, and the sulci were black in the T1W images. The grey matter had lower signal intensity than the white matter. In the more caudal slices the cerebrum and the brain stem were clearly visible. Bone marrow was bright white and the muscles grey as in sagittal T1W images. FIG.6.A,B,C.

**T2W images of normal dogs**

In the T2W images the grey matter was brighter than the white matter. CSF in the ventricular system and on the surface of the brain had high signal intensity and appeared white. Cortical bone and air appeared black with very low signal intensity. Fat had high signal intensity and appeared bright white. FIG.7.

![FIG.7. T2W transverse image of a normal pointer (N3) with asymmetry of the lateral ventricles as in Fig 1. The CSF is white and the white matter is darker than grey matter. 1.0T, SE2 TR 2200/TE 80, FOV 230, matrix 256x256. Imaging time 9 min 26 sec.](image)

**Normal proton density images**

Grey and white matter had intermediate signal intensity. White matter was darker than grey matter. Cerebrospinal fluid was dark in the PDW images thought its signal intensity was slightly higher than in T1W images. Fat had high signal intensity. Air and cortical bone were black because of low signal intensity FIG.8.

The usefulness of these differences in tissues created by different sequences was illustrated in the case studies.

![FIG.6. A-C. T1W images of the same normal pointer N3 with asymmetry of the lateral ventricles. A: Transverse T1W image. Field strength 0.1T, PS 3D TR 50/TE 20, matrix 192x256, FOV 192x256, slice thickness 4 mm scan time 10 min 14 sec, (lateral ventricle arrows). B Transverse T1W SE image. Field strength 1.0T, slice thickness 5 mm, TR 600/TE 15, matrix 256x256, FOV 230x230, scan time 5 min 10 sec. C: T1W sagittal image. Field strength 0.1T, PS 3D TR 50/TE 20, FA 60°, matrix 256x256, FOV 307x307, slice thickness 5mm.](image)
ULTRALOW FIELD (0.02T, 0.04T) IMAGING

Normal dogs

The normal appearance of tissues was determined from scans of two dogs: an English pointer (N2) was scanned with 0.02T imager only in the sagittal plane, and an English setter (N4) with the 0.04T imager, which was reported in publication I (Publication I, Fig 1A, B).

MRI of the brain of patients

The following features in images were noted: location of pathological changes, mass effect, ventricular size, signal intensity, signs suggesting presence of oedema, and effect of contrast enhancement. The visibility of these in images created by the different machines was noted.

ULTRALOW FIELD (0.02T, 0.04T) IMAGING

Developmental anomalies including unexplained hydrocephalus

Cases not previously reported

Anatomy of the ventricles

The ventricles were seen well because of the fluid filling with low signal intensity (black) in T1W images and with high signal intensity (white) in T2W images. In proton density images the anatomy was not so distinct as the signal intensity of the fluid was of intermediate signal intensity (grey) rather similar to the signal intensity of the white matter. In the transverse sections the lateral ventricles were symmetrical in shape and size in all normal dogs but one (N3), where the left lateral ventricle was three times larger than the right one, both in 0.1T and 1.0T images (publication IV Fig. 1. C-D). The lateral ventricles could also been evaluated from the dorsal views. The third ventricle could be seen in normal dogs best in transverse sections. The fourth ventricle was seen best in midsagittal and dorsal views.

P10, a collie pup with a behavioural disorder had larger lateral ventricles than the normal littermate, P11, in images made with 0.04T imager. We could not evaluate the significance of this finding because we have not imaged enough healthy collies to compare. Though collie is not one of the breeds described in the literature as predisposed for large ventricles. Small findings in the brain of this dog could not be excluded because of the poor spatial resolution due to the slice thickness (10 mm in sagittal and 5 mm in transverse images). The dogs were lost from follow up.

P20, a Japanese chin with neurological signs of hydrocephalus had the diagnosis confirmed in the 0.04T T1W transverse scan. The brain of this dog was so small that details of the anatomy like gyri were not visible as the coil was relatively too large and the head could not be positioned in the middle of it. The images were too grainy to see details of anatomy of the brain of this 3 kg dog: only the very large dark lateral ventricles and the very thin brain
cortex were clearly seen. The sagittal T2W scan had overall a too high signal intensity that obscured anatomical details of the brain. Necropsy verified a hydrocephalus for which no cause was found.

**Infection**

*Cases not previously reported*

P4, a Griffon Bruxellois underwent first an ultralow field strength (0.04T) and later a high field-strength (1.0T) imaging. In the first MRI with 0.04T scanner a mixed signal intensity mass with a diffuse outline was located in the thalamus and in the basal ganglia region. Both lateral ventricles were large and asymmetric, FIG.9.A. Five months later the high field MRI revealed the same findings in the T1W images, FIG.9.B. The lesion was an extensive mass with mixed signal intensity in the midline and on its right side there was also an area of mixed intensity that indicated a lesion in the white matter. Both lateral ventricles were even more dilated than in previous images. Contrast studies or T2W images were not made because of the dog’s poor condition. The differential diagnosis between tumour and abscess was not possible neither in ultralow- nor in the high field images. Necropsy showed a brain abscess.

*FIG.9.A. Transverse-oblique section of brain of a Griffon Bruxellois (P4) with brain abscess. A space occupying mixed signal intensity mass (arrow) and asymmetrical large ventricles can be seen in both T1W images. Fig. A: A low field 0.1T image (PS 3D TE 100/ TR 90). The deviation of midline of brain (arrow) is seen in this image better than in Fig.9.B.*

**Degenerative conditions**

*Previously reported*

Ceroid lipofuscinosis

P6, an English setter (publication I, Fig. 3.) with histologically verified ceroid lipofuscinosis of brain showed in T1W 0.04T images decreased brain mass with abnormally wide sulci and dilatation of all the ventricles. The signal intensity of CSF around the brain substance was as low as in the ventricles. The black CSF could not be distinguished from the signal void of the cortical bone density. A normal dog (C4) of the same breed was used for comparison. It was imaged with the same sequence with the same machine (publication I, Fig 2.). With T1 weighting the cortical bone and CSF could not be differentiated.

*Cases not previously reported*

Encephalomalacia

An eight-year-old Irish terrier (P7) that had had epileptic seizures and progressive loss of vision in the last 4 years was imaged with 0.04T imager. The T1W images without or with contrast did not reveal abnormalities. The images had poor resolution and disseminated lesions caused by histologically verified encephalomalacia were missed.

*FIG.9.B. Fig. B. A high field 1.0T image (SE TR 2500/ TE 25). The deviation of midline is not so distinct but the anatomical and pathological details in the brain are better seen than in Fig.9 A of the same patient.*
Cerebellar atrophy

P8 and P9, two Kerry Blue terriers with histologically verified cerebellar atrophy had no pathologic findings in 0.04T images. The pulse sequence for T1W transverse scans showed anatomy well, only the distinction of sulci was not clear. The T2W sagittal images had too high signal intensity for evaluation of details other than the outline of the brain, similar to P20. The necropsy verified a cerebellar atrophy. The reduction in size of the cerebellum was not noticeable macroscopically.

LOW FIELD STRENGTH (0.1T) IMAGING

Normal dogs

PREVIOUSLY REPORTED

N3, a pointer was scanned with low field (0.1T) and high field strength (1.0T) units. Both gave good anatomic images of the brain. The IR sequence showed better than the corresponding image with 1.0T (SE TR 600/TE 15) the differentiation of grey and white matter with good contrast resolution. The 1.0T images had a good spatial resolution giving a more pleasing appearance, but did not give more information. (Publication IV, Fig 1. A-D).

N5, N9-10 Beagles with normal brain in 0.1T images were reported in publication V. (publication V Fig.3. A-C). The dog’s images are described more in detail in the section of New Coil Design.

Patients

Seizures

PREVIOUSLY REPORTED

A three-year-old German shepherd (P18) had had severe seizures since three months of age. The dog was first scanned with a low field strength scanner and is described in publication IV Fig 3. A-D. It was interpreted as a frontothemoidal meningocele, as it had MRI signs very similar to a case in a human patient (Naidich et al, 1987). The application of contrast did not show any additional findings. The T2W images showed the overall anatomy less distinctly than the T1W images but the fluid in the ventricles was clearly seen because of its high signal intensity, FIG. 10.B. A year later this dog was scanned with a high field unit. The T1W high field images showed the same findings as the earlier low field images, FIG.10.A,C. Additionally a narrow, high signal intensity rim delineating the lateral ventricles on the T1W contrast images was seen. Similar changes were described in human medicine as transependymal spread of CSF out of the ventricles under a pressure gradient (Russell 1949). The distinction of the sulci and the fine details of the changes were better in high field strength images than in low field images. This did not add to the diagnosis made from the low field strength scans. The dog was not available for necropsy as it survived another four years.

FIG.10.A. Frontothemoidal meningocele in P18. 0.1T: T1W sagittal image of brain, 3D GRE TR 50/TE 15, matrix 256x256, FOV 256x256, scan time 6 min 30 sec, slice thickness 5 mm. The areas with CSF are black (as is also the air filled sinus frontalis). The local thinning of frontal bone is seen as the loss of medullary fat (no high signal intensity) in that area of the bone.
FIG. 10.B. 0.1T T2W sagittal image, FSE TR 2300/TE 120, matrix 192x256, FOV 250x330, imaging time 7min 22 sec. The heavily T2W image does not show the overall anatomy well. The lateral ventricles have high signal intensity (white) as is also the fat in the bone marrow. Notice the differences in signal intensity of CSF. It is less intensive (grey, arrow) in areas of more rapid flow than in areas of less rapid flow for example in the nasal bulgings because of the difference in the saturation of fluid.

FIG. 10.C. 1.0T T1W sagittal image, SE TR 330/TE 15, matrix 256x256, FOV 250x250, scan time 2 min 28 sec slice thickness 5 mm. The areas with CSF are black as is also the air filled sinus frontalis. The local thinning of frontal bone is seen as the loss of medullary fat (no high signal intensity) in that area of the bone.

P24-29, six Newfoundland pups with severe seizures and neurological signs of hydrocephalus, all had varying degrees of lateral ventricular enlargement and brain atrophy symmetrically or asymmetrically. These were best evaluated with T1W dorsal scans, FIG 12.A. The transverse scan gave also a good anatomic image, FIG 12. B,C. The sagittal T1W scan showed the anatomy and the pathological changes well in these sections FIG 12. D,E. Contrast application did not give any additional findings. The T2W and PDW in transverse direction showed the CSF well. These malformations were verified at necropsy. The etiology remained obscure.

P31 (publication IV, Figure 2. A,B), a German shepherd was scanned with low- and high field strength scanners because of a change in behaviour. Very large symmetrical lateral ventricles and loss of the septum were seen in both examinations, FIG.11. Contrast application did not give any additional information. In the dorsal scan the small FOV caused a “wraparound” artifact as the nasal part showed on the actual image but fortunately not on the brain area. The imaging times of the tranverse scans were four times longer in the low field scanner than in the high field imager. The necropsy verified hydrocephalus but not its cause.
**FIG.12.A.** Dorsal 0.1T TIW scan of a Newfoundland pup (P29). The low signal intensity (black) lateral ventricles are asymmetrically dilated. The septum is incomplete (arrow). PS 3D TE 60/TR 25, FA 70°, matrix 256x256, FOV 230x230, imaging time 8 min 52 sec.

**FIG.12.B.** A transverse 0.1T PDW scan of the brain of a Newfoundland pup, P24. The CSF in the dilated ventricles is seen with intermediate signal intensity (greyish). DE TR 2500/TE 30, FA 90°, matrix 256x256, FOV 256x256, slice thickness 5 mm, scan time 10 min 40 sec.

**FIG.12.C.** A transverse 0.1T TIW scan of the brain of a Newfoundland pup, P25 (a littermate of P24, with maximally dilated, low signal intensity (black) ventricles, and cortical atrophy.

**FIG.12.D-E.** The same Newfoundland pup, P24, as in FIG.12.B. Two different parasagittal TIW scans, PS 3D TE 69/TR 15, FA 60°, matrix 256x256, FOV 256x256, slice thickness 5 mm, scan time 4 min 6 sec. In FIG.12.D, the thinning of the brain cortex (arrow) and the bulging of the lateral ventricle are seen with the low signal intensity (black) CSF. In FIG.12.E, In the caudal brain area dorsally in the cerebellum, a region of lower signal intensity (arrow), is most likely caused by partial volume averaging of the low signal intensity CSF and of the intermediate signal intensity brain substance, creating an artifact.

P32, a young St Bernard dog was referred for MRI because of severe seizures. The transverse and sagittal T1W sequences gave good anatomical images of the very large lateral ventricles with typical low signal intensity. The T1W images after contrast application did not show any enhancement. The result from necropsy was hydrocephalus of unknown cause.

**Miscellaneous neurological signs**

**CASES NOT PREVIOUSLY REPORTED**

P21, a Rottweiler with change in behaviour was imaged with 0.1T in sagittal and transverse scans. The brain anatomy was well visualized. A T2W fast spin echo pulse sequence image was grainy but the ventricles were well visible. One of the
lateral ventricles was markedly larger than the other one. No enhancement was seen with contrast. The necropsy revealed no cause for the asymmetrical ventricles. The pathologist interpreted this as a normal variation in ventricle size.

P22, another Rottweiler with ataxia was imaged with 0.1T in transverse, sagittal and dorsal sections with T1 and T2 weighting. The anatomy was well seen in the images. The T1W images had better spatial resolution than the T2W ones. The transverse images showed asymmetry of the lateral ventricles but this could be a normal variation in the ventricle size. A contrast brain scan did not add to the findings. The additional cervical myelography and MRI showed a slight ventral compression of the spinal cord in C2-C3 area. The dog was euthanized because of the cervical spinal ataxia. The owner refused necropsy.

P34, a mongrel with episodes of balance problems and apathy had transverse images made with the 0.1T scanner. The T1W, T2W and PDW sequences showed good anatomic images and sufficient spatial resolution but no specific MR findings. The owner wanted to keep the dog. It was lost from follow up.

**High field strength (1.0T) imaging**

**Normal dogs**

The images of the brain of a normal pointer (N3) are shown in publication IV, Fig 1. B,D. The description of the visibility of different structures in each imaging are described at the beginning of this chapter. A normal German pointer (N1) had similar findings like N3 but symmetrical lateral ventricles in 1.0T imaging.

**Patients**

**HORNER’S SYNDROME**

P12, a German pointer, with signs of Horner’s syndrome had normal brain anatomy in the images both at 7 years of age in ultralow- (0.04T) and at 8 years of age in high field strength MRI. The unilateral masticatory muscle atrophy and the clinical signs persisted but the condition was tolerable with symptomatic treatment.

**Brain mass, infectious**

**CASES NOT PREVIOUSLY REPORTED**

P4, a Griffon Bruxelloise, has already been described under low field strength imaging as its first examination. Five months later high field MRI revealed the same finding. The lesion was extensive and in the midline and to its right side there was an area of mixed signal intensity. Both lateral ventricles were even more dilated than in previous images. The differential diagnosis between tumour and abscess was not possible even in these images. The better resolution of the high field strength images did not add to the findings seen with ultralow field strength imaging. FIG.9. A,B. Necropsy verified a brain abscess.

P30, a Poodle, with central vestibular syndrome had high titer of borreliosis in CSF. It had MRI to rule out brain tumour. In T2W images a lesion on the left side in the brainstem was seen as an area of mixed signal intensity with increased signal intensity borders as sign of oedema. These areas were best seen in dorsal views in T2W images. The contrast studies in sagittal and transverse views with T1 weighting showed an enhancement in the same area and several focal spots of enhancement in various locations in the white matter. The symptoms disappeared after treatment with antibiotics for borreliosis. The dog was alive two years after these examinations and lost for further follow up.

**Brain mass, neoplastic**

**CASES NOT PREVIOUSLY REPORTED**

P16, a Miniature schnauzer, with severe cerebral disorder and suspicion of a brain mass, showed a lesion in brainstem in high field imaging. An oval shaped low signal intensity area with diffuse margins in the middle of brainstem was seen in the T1W transverse-oblique images, FIG.13. A. The T2W transverse images showed a high signal intensity area as a sign of oedema around a less intensive inner part of the actual mass through the whole brainstem, FIG.13. B. Because of the poor
condition of the dog neither other planes nor contrast studies were done. The transverse-oblique plane was chosen to save time as no separate transverse and dorsal images were then necessary. The brainstem mass lesion was a histologically verified astrocytoma.

**FIG.13.A.** Miniature schnauzer (P16) with a histologically verified brainstem astrocytoma. **A.** 1.0T T1W transverse-oblique scan of the brain. The lesion in the brainstem is shown as a lower signal intensity area compared to the surrounding brain tissue (arrow). TR 600/TE 15, matrix 256x256, FOV 220x220, scan time 5 min 10 sec.

**FIG.13.B.** 1.0T T2W transverse scan of a brain. There is a region with irregular signal intensity in the central part of the brainstem. The area, which shows lower signal intensity than the surrounding brain tissue is the tumour (arrow). The high signal intensity in the brainstem is from fluid (oedema). Note the lateral ventricles (fluid) also with high signal intensity in T2W images. TE 2500/TR 90, matrix 256x256, FOV 220x220, slice thickness 5 mm, scan time 10 min 44 sec.

**FIG.14.** 1.0T T2W transverse-oblique scan of the brain of the dachshund (P5) with malignant ependymoma. The brain mass (lower arrow) is seen as an area of mixed signal intensity in the pons. Large areas of white matter (normally dark in T2W image) have high signal intensity (white) as a sign of spreading white matter oedema (upper arrow). SE TR 2500/TE 90, matrix 256x256, FOV 200x200, slice thickness 4 mm, scan time 10 min 44 s.
Cerebellar atrophy

Cases not previously reported

P33, a 6-month-old Finnish hound had shown progressive signs typical of cerebellar ataxia. The cerebellum was very small and was seen best in the mid-sagittal images of the T1W sequence (FIG.15. A.). The transverse and dorsal sections with T1 weighting showed the anatomy well but the evaluation of the size of the cerebellum was not so clear as in the sagittal images. The transverse T2W and the PDW images did not add to the findings. The macroscopic pathology showed a very small cerebellum.

FIG.15.A. Finnish hound (P33) with cerebellar ataxia. 1.0T T1W sagittal scan of the brain. The cerebellum is relatively small (arrow). SE TR 350/TE 15, slice thickness 4 mm, scan time 2 min 28 sec.

Imaging of the spine

MRI characteristics of tissues in the spine with all the used field strengths

T1W images

T1W images offered the best anatomical image with all the different field strength scanners. The muscles showed varying shades of grey with darker stripes of fibrous tissue. The osseous structures appeared black as areas of relative signal void. The bone marrow with fat was of high signal intensity (white). The CSF, medulla and epidural fat could not be distinguished well from each other and appeared as various shades of grey. The normal disks were homogenous grey (intermediate signal intensity). FIG.16.

FIG.16. The lumbosacral spine of a normal pointer (N3) in a low field 0.1T scan. The epidural fat has high signal intensity (white) and shows no compressions of spinal cord (grey, arrow). The disks have intermediate signal intensity (grey) in T1W images. PS TE 155/TR 20, FA 90°, FOV 302x358, matrix 216x256, scan time 4 min 20 sec.

T2W images

On T2W images the signal of cancellous bone was diminished and the distinction between cortical and cancellous bone was not so clear as in T1W images. In T2W images the nucleus of a normal disk appeared as high-signal intensity bright (white) because of its higher water content, and the annulus fibrosus appeared dark as a low-signal intensity. The T2W image produced a myelographic effect as the CSF and epidural fat appeared bright white, due to its higher signal intensity, around the mid-range signal intensity of the spinal cord (N2, Publication I, Fig.2). In the T2W high field strength scan the grey matter was brighter than the white matter of the medulla.

Proton density weighted (PDW) images

The signal intensities of different tissues in a PDW image had less contrast between them than in T1W or T2W images. The cortical bone was black and cancellous bone slightly brighter. The more signal intensive CSF could be better distinguished from the less intense medulla but the nucleus pulposus did not stand out as clearly as in the corresponding T2W images of the same normal dog (N6) imaged with the 1.0T high field scanner. FIG.17.
**FIG.17.** 1.0T PDW sagittal scan of sacrolumbar spine in the normal Rottweiler, N6. The disks and CSF are of intermediate signal intensity (arrows). Fat around the spinal cord has high signal intensity. The cortical bone has low signal intensity (black). SE TR 2500/TE 15, matrix 200x256, FOV 260x260, slice thickness 4 mm, scan time 8 min 24 sec.

**FIG.19.** 1.0T PDW transverse scan of the cervical spinal cord. The grey substance appears with brighter signal intensity and the white substance with lower signal intensity. SE TR 2500/TE 25, slice thickness 5 mm, scan time 10 min 44 sec.

**TRANSVERSE IMAGES GENERALLY**

The transverse planes with low field strength scans could give only a view of the diameter of the medulla and the amount of CSF and fat around it. **FIG.18.**

**FIG.18.** An ultralow field (0.04T) T1W transverse scan of a German shepherd’s (P52) lumbar spine. The spinal cord (arrow) has higher signal intensity than the CSF around it but no details inside it can be seen. Fat has high signal intensity (white). PS TE 125/TR 40, slice thickness 10 mm.

Dorsal planes of the spine

Dorsal planes of the spine were used for exact localization of a spinal tumour in the bullmastiff, P45. The extent and laterality of the tumour could so be better visualized for surgery with the addition of this plane.

**Effect of the size of field-of-view on image quality**

With the 0.1T scanner, which had variable FOV, using a small FOV did not improve the image quality sufficiently to justify the additional time required to make multiple images with the small FOV. Even the largest FOV (358x358 with 256x256 matrix) gave good detail in the middle of the image but the peripheral parts were less distinct. We used a large FOV because we wanted to evaluate the whole cervical, thoracic or lumbar area in as short time as possible (patient P43).

Normal dogs

In the spine the contrast between the grey and the white matter, the fat surrounding the cord, the bone and the disks are all of importance in evaluating an area for lesions. We compared the sequences available on each machine for their capacity to enhance tissue contrasts. The use of these different sequences is described in the case studies. The description is based on the normal

The anatomy of the medulla with distinction of the white and grey matter was seen in the transverse scans of cervical spine with high field MRI. **FIG.19.**
dogs (publication III, Table 1.) and patient findings.

**ULTRA LOW FIELD STRENGTH EXAMINATIONS (0.02T AND 0.04T)**

Normal dogs

In the ultralow field examinations of normal dogs' lumbar spine with T1 weighting, the cortical bone and ligaments had low signal intensity. Spinal cord and intervertebral disks had intermediate signal intensity. Bone marrow and epidural fat had high signal intensity. In T2W images the low signal areas were the same as in T1W images. High signal intensity was seen from disks, CSF and bone marrow. The vertebrae and soft tissues had intermediate signal intensity and appeared as various shades of grey. The epidural fat in the vertebral canal had high signal intensity. (Publication I, Fig 2, and publication III, Fig. 1. A-B).

Patients

Cervical cord compression (wobbler syndrome)

*Cases not previously reported*

Four Rottweilers (P22, P39, P40) were examined with 0.04T scanner. P22 and P39 had only T2W scans, as the disks were the main interest. It was not possible to extend the cervical spine of P22 sufficiently to get a definite diagnosis, although the cord at C2-3 looked slightly compressed, as it had been on the myelogram. P39 had slightly decreased signal intensity of disk C6-7 in the sagittal scan. This disk appeared to protrude into the spinal canal and may have compressed the cord, but the neck could not be sufficiently extended to get a good sagittal scan. Neither of these two dogs was operated due to the indeterminate diagnosis. P40, another Rottweiler with similar clinical signs, had a myelogram that suggested swelling of the spinal cord at the level of C5. With 0.04T, the sagittal T1W images showed normal intermediate signal intensity of the disks and no compression of the cord. This sagittal view was better than in P39 as the stretching out of the neck was better. The T2W scans were not as good because they were very noisy. The disks could however be seen with high signal intensity. No cord compression was seen, and no cause of the apparent obstruction. Necropsy revealed vascular changes in the cord in the area of C5 related to a chronic compression. The resolution of the 0.04T images was not good enough to detect small lesions like vascular changes as the slice thickness was 10 mm.

P41, with clinical signs of cervical spinal cord compression was imaged with the same 0.04T scanner in sagittal plane with T1W and T2W sequences. The images of the disks did not show compression.

Trauma

P35, a Finnish hound hit by car had quadriplegia. It was scanned with the 0.02T scanner. The sagittal T1W images showed small irregular areas with decreased signal intensity that indicated cord damage (increase of fluid, possibly bleeding) at the level of C2, FIG.20. The necropsy findings of haematoma around the spinal cord and a fracture of the C2 could not be seen because the signals from both, the bone and chronic haematoma are very low. The low spatial resolution of the 0.02T scanner made it impossible to differentiate them.

*FIG.20. A sagittal scan with ultralow field 0.02T imager of the cervical spine of the Finnish hound P35. The cortical bone appears with no signals (black in the image), no details of bone structure can be seen. Spinal cord (arrow). PS TE 1500/TR 80, slice thickness 9 mm.*
Disk degeneration and prolapse including lumbosacral compression

**PREVIOUSLY REPORTED CASES**

P36, a German shepherd is described in Publication I (Fig. 4). In the T2W images the intervertebral space between C2 and C3 gave no signals. A non-signal mass was seen compressing the medulla ventrally in the same area. The diagnosis of disk prolapse at this site was confirmed at surgery.

Publication III: For lumbar spine evaluation for degenerative diseases of spine eleven patients (P46-56) were examined with 0.04 T scanner. They had neurological signs referable to the lumbar area clinically, neurologically and radiographically. Disk degenerations and prolapses including lumbosacral compression were found in MRI (Publication III, Figures 1-5). These dogs and their findings are described in the publication III (Figures 1-5).

In patients P46-50 (in publ. III: Patients 1-5), a lumbosacral stenosis was suspected on the basis of neurologic examination and radiographs. In the MR images one of these dogs had total signal loss and four had reduced signal intensity at the L7-S1 space. All dogs had a low signal area around the endplates, and the vertebral canal was narrowed at the lumbosacral junction. The best pulse sequence used with the 0.04T scanner for sagittal T1W images was PS 400/40, and for T2W images especially for showing the disks was PS 1000/60. The transverse images were made with PS 125/40, 10 mm slice thickness. These transverse views did not help in the diagnosis as the cord was too small to be seen even in large dogs like German shepherds. A compression of the cauda equina was diagnosed on the sagittal scans. The dogs were euthanized because of the severe clinical signs and the confirmation of diagnosis with MRI.

The patients P51-55 had a suspected lumbar disk disease (in publication III, patients 6-10). Four of them had degeneration of one or more disks. One had additionally mild spondylosis. Two had normal disks, with some evidence of spondylosis of the caudal lumbar spine. The changes were seen in both, radiographs and MRI.

**Low field examinations (0.1T)**

**Trauma**

**CASES NOT PREVIOUSLY REPORTED**

P44, a Belgian shepherd with post-traumatic hyperextension of the forelimbs, opisthotonos and inability to stand up was referred for MRI. Previously done myelography had shown a filling defect of the uncertain origin and significance in the region of the 6th to 9th thoracic vertebra. The 0.1T MRIs scans showed no abnormal findings in the thoracic spine. The prognosis was therefore considered good.

Focal demyelination

**CASE NOT PREVIOUSLY REPORTED**

P43, an Australian cattle dog was referred for MRI because of progressive weakness in hind legs. In the T1W images the cord and the disks had similar intermediate signal intensity. CSF appeared with less signal intensity than the cord (that is normal). But in the region of the 10th thoracic vertebra on the sagittal view the cord appeared to disappear. It was visible on the tranverse section but was very narrow. The apparent disappearance of spinal cord structure was most probably caused by the signals from the increased CSF volume around the thin cord averaged with those of the cord in the same voxel (PVA), FIG.21.A. Transverse, sagittal and dorsal images were made with pulse sequence CBASS 3D TR 15/TE7.5 because of its “myelographic effect”. These transverse sections showed the lesion as a thinning of the low signal intensity cord up to a third of its normal thickness. A wide zone of high signal intensity around the cord was due to an accumulation of CSF in the wide subarachnoid space and the fat. There was no compression of the cord. Here the CSF is seen bright white, surrounding the less signal intensive medulla, providing high contrast, FIG.21.A. The T1W scan after contrast application showed no enhancement. All the three imaging planes were made to determine if an operation would benefit the dog. It was concluded to be inoperable.
Necropsy revealed a demyelination of the cord at the level of the 10th thoracic vertebra.

**High field strength (1.0T) examinations**

Normal dogs

The signal intensities of different spinal structures are the same in the whole spine of dogs.

Disk degeneration or prolapse including suspicion of lumbosacral compression

**CASE NOT PREVIOUSLY REPORTED**

P38, a five-year-old male German shepherd had shown pain without neurological deficits during attack-tests. A faint mineralization of the intervertebral disk between the fifth and the sixth cervical vertebrae was present in plain radiographs. This lesion could not be seen in the high field images as all the disks had the bright signal intensity of a normal disk. There were no other findings. The diagnosis from MRI was normal cervical spine. The dog managed well some years with physiotherapy.

**FIG.21.A.** The Australian cattle dog, P43 with local demyelination of the spinal cord. 0.1T T2W transverse scan at the level of the 10th thoracic vertebra. CSF and epidural fat give bright contrast to the low signal intensity spinal cord (arrow) that is thin at this section. The myelographic effect of the sequence CBASS 3D TR 16/TE 8, FA 90°, matrix 256x256, FOV 256x256, slice thickness 7 min 45 sec, is well demonstrated.

B

**FIG.21.B.** The Australian cattle dog, P43 with local demyelination of the spinal cord. B. 0.1T T1W sagittal scan. The CSF has lower signal intensity than the spinal cord, which is normal. The spinal cord has less intensive signal appearance in the 10th thoracic vertebral region than in other regions (arrow).

Neoplasia

**CASE NOT PREVIOUSLY REPORTED**

P45, a bull mastiff was found to have an expansive tumour mass in the spinal canal in thoracic area in MRI. The imaging was made in three directions for planning of surgery. In the sagittal slices of T1W images a mass was seen with decreased signal intensity compared to the cord and it seemed to spread out from the spinal canal. The destruction of the surrounding bone could be seen. The transverse images showed the decreased intensity mass clearly, filling almost the whole canal and compressing the cord severely. Dorsal scans confirmed these findings. The following operation and necropsy showed the same spread of the neoplasm. The histologic diagnosis was glioblastoma multiforme. FIG.22.A.B.

**FIG.22.A.** Sagittal T1W scan of the thoracic spine of the bullmastiff P43 with glioblastoma multiforme in the spinal canal. The tumour appears as dark grey low signal intensity area (arrow). The epidural fat and the CSF are bright with high signal intensity TR 600/TE 12.0, FOV 250x400 matrix 256x256, slice thickness 3 mm, scan time 3 min 9 sec.
symptomatically under regular veterinary control as long as it was comfortable for the dog. The owners found the remaining four years their pet could spend with them very precious because of this dog’s outstanding character during its whole life.

Patient P6, a young English setter, was euthanized because a brain atrophy seen on MRI confirmed the clinical suspicion of neuronal ceroid lipofuscinosis. This was verified in the histological examination.

Three patients, a Japanese Chin (P20), a German shepherd (P31), and a St. Bernard dog (P32) had very large symmetric ventricles. The owners wanted to know MRI results before making decision of euthanasia. As their clinical signs could not be controlled with medication the dogs were euthanized. The reason for the large ventricles was not found in necropsy.

The Newfoundland littermate pups (P24-29) with varying degrees of brain malformations were all euthanized. Before MRI was done, all kinds of treatments in different clinics were tried out but the clinical signs worsened. The owners could find some relief in their distress to decide upon euthanasia as they could understand after MRI that the pups could not be cured. The necropsy showed a variety of brain malformations with very large ventricles and cortical atrophy. The etiology remained unclear. In the pathologic examination toxoplasmosis was mentioned as one possibility.

The finding of large brain masses, most likely tumours (4 histologically verified) in MRI of 5 dogs (P1-3, 5) allowed the owners a somewhat easier decision for euthanasia. The owner of the miniature schnauzer (P16) wanted to keep the dog as long as possible by using symptomatic treatment. It died later at home. At necropsy the finding was a brainstem astrocytoma. In a Griffon Bruxellois (P4) the large mass lesion compressing one lateral ventricle was seen to expand between two imagings with 5 months interval. Although only the differential diagnosis between a brain abscess and tumour could be made, the dog was treated with antibiotics and corticosteroids. After the second imaging it was decided not to let the dog suffer any more.

Of the three patients with cerebellar ataxia, the two Kerry Blue terriers (P8, P9) did not show a

THE NEW BRAIN COIL FOR LOW FIELD STRENGTH SCANNER

Three normal Beagles were imaged with a dedicated Beagle brain coil, and for comparison a standard knee coil, and a MPXS coil (flexible multipurpose coil). The SNR comparison of the coils showed an improved SNR with the Beagle brain coil of more than 20%, as compared with that of the flexible coil, and 70%, as compared with the standard knee coil supplied by the manufacturer (Publication V).

THE EFFECT OF THE MRI DIAGNOSIS ON TREATMENT AND OWNERS’ DECISIONS

MRI of brain

Seizures were the most usual (18 patients: P13-15, 17, 18, 19, 21-23, 24-29, 32) reason for brain MRI. In seven (P13-15, 17, 19, 21, 23) of these cases the brain image was normal and the patients were treated symptomatically, as the owners were happy to keep their dogs. In the patients with seizures with normal MR images the exclusion of a life threatening mass was helpful to the owners and the veterinarian to decide to begin or to continue a symptomatic medical treatment. The German Shepherd, P18, with a congenital brain defect, a frontal meningocele, was imaged one year later to see if there was any change. As the situation was unchanged according to the MRI examinations the owners wanted to continue to treat the dog

FIG. 22. B. Dorsal TIW scan of the same area. The tumour appears as dark grey low signal intensity area (arrow). The epidural fat and the CSF are bright with high signal intensity TR 600/TE 12.0, FOV 250x400 matrix 256x256, slice thickness 3 mm, scan time 3 min 9 sec
relatively small cerebellum on MRI, but the Finnish hound’s (P33) cerebellum was clearly smaller than average. All three were euthanized because of the clinical signs typical of this progressive disease. The MRI made no difference in these decisions. The necropsy verified cerebellar atrophy in all three.

The brain of the Rottweiler (P22) with ataxia had a normal brain anatomy on the MRI images but suspected cord compression at C2-3. The owner took the dog home and it was lost from follow up.

One of the two collie littermates (P10-11) with cerebral disorders, P10, with more pronounced clinical signs had very large asymmetric lateral ventricles and the other’s brain image was normal. The owner did not make any decision after MRI at that time and they were lost from follow up.

A Poodle (P30) had central vestibular neurological signs because of borreliosis. MRI was made to try to rule out a tumour. The findings did not allow that small tumours or other lesions which could cause the BBB to be disrupted could be excluded. The dog was treated after diagnosis with antibiotics specifically against this infection. It recovered gradually and the owners were happy to keep it.

A German pointer (P12) with Horner’s syndrome had normal brain anatomy in the images both at 7 years of age in ultralow- and at high field strength MRI at 8 years of age. The unilateral masticatory muscle atrophy and the clinical signs persisted but the condition was tolerable with symptomatic treatment. The owner let his hunting dog live in retirement as a pet as no brain lesions were detected in MRI.

The mongrel (P34) with a cerebral disorder without any etiological clinical diagnosis had normal brain on MRI. The owner decided to keep the dog and returned to the referring veterinarian for symptomatic treatment and control.

**MRI of spine**

The MRI with 0.02T imager did not show the fracture and haematoma in a Finnish hound (P35). The dog was euthanized because of quadriplegia, rather than the MRI results.

The German shepherd (P36, publication I, Fig.4) with lameness and neck pain was operated because of a disk prolapse seen on MRI. The removal of the prolapsed disk material was succesful and the dog recovered well.

The two German shepherds (P37-38) with neck pain had no abnormal findings in MRI. Both were kept as active working dogs and were treated symptomatically. The other competed successfully as an attack dog on national and international level several years.

The “wobbler”-syndrome patients (P22, 39-42) were all Rottweilers. P22 was euthanized. The owner did not want any necropsy done. In P39 the reason for cervical spinal ataxia was determined to be a degenerated protruding disk at C6-7 by radiography and MRI. The owner chose euthanasia because of this finding and earlier diagnosed osteochondritis dissecans and arthropis in elbow joints. The owner of P40 decided to let the dog be euthanized because of the severe clinical signs and swelling of the cord in the myelogram, not because of the findings in MRI. P41 with severe signs but no MRI evidence of cervical spinal compression was lost from follow up as the owner took it home. In P42 the spinal compression visible in the flexed views on the myelogram could not be seen in the MRI extended views. Because the myelogram showed a dynamic compression the MRI was considered a false negative. The necropsy confirmed the compression.

The Australian cattle dog (P43) with an “hour glass” narrowing of the medulla in the Th10 area and no compression was euthanized as the MRI showed that the dog could not be helped by surgery, nor by symptomatic treatment, which had been tried earlier.

The Belgian shepherd (P44) scanned after an accident was given a guarded prognosis because of the severe signs in the first days. As the MR images did not reveal serious lesion the owners were encouraged to persevere. The dog recovered with symptomatic treatment after some days.

The bull mastiff (P45) had an invasive mass spreading out from the medulla to the spinal canal and to adjacent tissues on MRI. In spite of the poor prognosis the owner wanted surgery. In agreement with the owner the dog was euthanized during the operation after the massive
tumour spread was verified. The necropsy finding was glioma in the medulla and adjacent tissues.

In patients P46-50 (in publ. III: Patients 1-5), a lumbosacral stenosis was suspected on the basis of neurologic examination and radiographs. The German shepherds owned by the Defence Forces (P46, 49-52) and the bull terrier (P48) were euthanized after MRI verification of the lesions. The privately owned giant schnauzer (P47) was euthanized as it had already once been operated for spinal stenosis and the signs were again increasing and MRI verified a compression of medulla.

The owners of the patients P53 (a dachshund) and 55 (a cocker spaniel) wanted to keep the dogs as MRI verified disk degeneration without protrusions.

Six out of the described seven Defence Forces owned German shepherds with neurological signs of compression of cauda equina verified by MRI, P46, 49-52, and P55, were euthanized. The seventh, a 10 year old the Defence Forces dog (P56), with no signs and no compression or other clinically significant MRI finding, retired from the Defence Forces service.
DISCUSSION

COMPARISON OF ULTRALOW-, LOW-, AND HIGH FIELD STRENGTH MRI IMAGERS

The case material used for the comparisons was heterogeneous, and the variables such as sequences used, depended on the knowledge at the time of use, and the constraints placed on the examination by the practicalities of patient management. The judgements are thus empirical, but based on the experience from 73 examinations with different machines.

We had no opportunity to use a mid field scanner.

There are many factors to be weighed when comparing equipment. Cost is of course very important in the veterinary profession, as it has to be borne directly by the animal owner, by research grants or by the Department involved, and not by a medical insurance plan, as in human medicine. In some cases, arrangements can be made to rent equipment used primarily for humans, in which latest case, the best quality can be expected. This discussion will not make judgements about cost, which varies with the individual circumstance, but compare the technical features of the machines and the image quality they provide based on the case material described.

The sequences that are used in different types of machines cannot directly be applied to other machines but are designed for each machine by physicists. A cooperation between a physicist and a veterinarian who knows what he or she can expect from the study enables a physicist (operator) to create appropriate sequences to get diagnostic MR images of dogs. Testing sequences to find suitable ones for the animal patients is very time consuming.

ULTRALOW FIELD STRENGTH

Technical advantages

Ultrasound field strength machines have a number of advantages. The cost of the machine, its installation and maintenance are relatively low. The physical design with a large short bore allows the dog to be examined, intravenous anaesthetic to be given without moving the dog, and easily repositioning for different views. It is quiet, which is better for the persons and for the dog. The low magnetic field is a low risk for flying metal objects and heating of metal implants such as stainless steel bone pins. In ultralow– and low field systems the relaxation times, especially T1, are shorter and the relative differences of T1 between different tissues are larger (Sepponen et al, 1985)

Technical disadvantages

In the ultralow field strength imagers there were only few standard sequences. The 0.02T had three and the 0.04T had four sequences. The physicist had to create the best values for dog imaging in each case. This means a lot of investment of time from him and the machine. Greater slice thickness, use of interslice gap, and larger pixel size can compensate for an ultralow field machine’s slow acquisition. The latter improves signal intensity but lowers resolution, as does a thicker slice. No rapid sequences are available. If there is a particular kind of lesion suspected from the clinical information, to save time a specific sequence that maximises its detection can be selected, for example, T1 weighted for hydrocephalus (or mass lesion with contrast), and T2 weighted for tumour and oedema identification. However, a wrong guess would lead to an inadequate examination.

In the spine, it is particularly important to clinically localize the spinal lesion because of long imaging times.

Motion artifacts are a particular problem with the long scanning times of ultralow field machines. An adequate sedation or preferably anesthesia, if available, and the positioning of the dog exactly in the coil in the manner that is physiologically optimal for the dog help prevent motion disturbance.

The machine was not capable of reconstructing other slice orientations from the data set (no 3-D
imaging). Other planes had to be obtained by repositioning the dog’s head or spine, which required new "scout" images. This was a particular disadvantage in the evaluation of disks. All this took considerable time (range 5-20 minutes).

The fixed FOV of the ultralow field machines was not in itself a disadvantage because the advantages of a variable FOV could not be used in these machines with their very limited computer capacity. A reduction of matrix size, had it been possible, would have reduced the imaging time.

Lack of a wide selection of coils limits resolution in images of small body parts. No special head coil for dogs or flexible coils were available and so close contact of the coil with dog head or spine was not possible. Thus the images were much noisier than those obtained with machines that had coils which fitted better to the head or spine of a dog regardless of field strength.

**Diagnostic quality**

**Brain**

The diagnostic quality of the ultralow field strength imager was found sufficient to show increased width of sulci of brain (publication I, Fig.3), with and without contrast application for brain masses (publication II, Figures 1-3). The T1 and T2W images with partial saturation pulse sequences gave a good contrast between brain tissue and CSF in the images of a normal dog brain. T1 weighted images were sufficient to show enlarged brain sulci in a dog with lipofuscinosis (publication I, Figures 1,3) These images showed similar enlargement of the ventricles and cerebral atrophy as the CT images of a dog with ceroid lipofuscinosis described by Franks et al, (1999). As we did not take T2 weighted images of ceroid lipofuscinosis as in human MRI (Autti et al, 1992) we could not evaluate if we might have seen, besides the cerebral atrophy, a high periventricular intensity and changes in the signal intensity of white matter.

Encephalomalacia in P7 could not be shown even in contrast studies probably because of the very grainy image.

**Spine**

The image quality of normal spine and a dog with disk prolapse (publication I, Fig. 2, 4, and publication III, Figures 1, 2, 3b, 4, 5) showed diagnostic quality with both T1 and T2W imaging in sagittal planes. In our canine patients MRI method proved to be more sensitive than radiography in detecting disk degeneration of non-mineralized disks (publication III.). The resolution was not sufficient to see lesions in the spinal cord, only external compressions. Even mild herniations of L7-S1 disks could be seen in sagittal images. It was particularly valuable for evaluation of lumbosacral stenosis. The MR imaging of the lumbar spine showed the degenerated and protruded disks and compressions of spine also in lumbosacral area. Spondylosis and/or hypertrophied ligaments with their similar low signal intensity in low and high field MRI could be identified as causes of compression.

The transverse images did not add to the diagnostic information because of the small size of the disks and spinal cord.

In the imaging of spine both weightings were also helpful to see the disks and their degeneration status, and spinal compressions. If both weightings are done the possibility to see all the possible lesions increases. Disk prolapse can be shown only with T2W images as CSF has high signal intensity creating a “myelographic effect” around the spinal cord. The T2W scans take longer than the T1W scans but are more sensitive to show disk degeneration than are T1W images. On the other hand, T1W images with bright signal from the epidural fat can also help to identify a spinal cord compression (publication III). The high signal intensity epidural fat contrasts with a low signal intensity degenerated disk or mineralized mass and thus can localize a compression.

**LOW FIELD STRENGTH**

**Technical advantages**

These machines were a big advance in computer design and power from the machines of the 1980s, yet still were not much more expensive. They are also quiet, and have a short bore, retaining the
possibility to easily examine, dose with sedative and move the dog. The image quality was much improved: the elimination of the interslice gap greatly improved the resolution, the slices and pixel size could be thinner, and the variable FOV allowed a shorter examination time and better resolution with a small FOV. If the FOV is too small for dog’s entire head an aliasing artifact can hide parts of the findings. An interslice gap could be used if needed to make the scan time shorter. So often also dorsal planes could be created in addition to the sagittal and transverse ones actually scanned.

There were many more sequences available, and they were much faster than previously. T1W images gave a better differentiation of grey and white matter when IR sequence was used in a normal dog (N3) than the high field study of the same normal dog (N3). This is a finding in accordance with the experiences in MRI of humans by Doyle et al, (1981), where IR is especially used for this differentiation because of its sensitivity to distinguish between grey and white matter.

The degree of homogeneity of the magnetic field is proportional to the field strength of the imaging magnet. Magnetic field inhomogeneities degrade imaging. We used mostly GRE sequence because of the experiences of Virolainen et al, (1993), who found at 0.1T when the field strength is homogenous, the SNR is better with GRE than with SE techniques. The use of GRE sequences also enables the three-dimensional imaging and manipulation of contrast with the flip angle. In our scans we used a flip angle of 50-90°. With partial (less than 90°) flip angles the imaging times can be reduced by two- or fourfold when lesions with elevated T1 values are being examined (Mills TC et al, 1987). GRE scannings require a relatively short echo time. At low field strength, the disadvantages and artifacts associated with GRE pulse sequence are not as disturbing as they are in high field strength imaging. The low field units have sequences to take advantage of low field physics and produce different tissue contrasts, particularly with T1W or gradient echo (T2*) images (Kim T et al, 1992).

The better transmitter and receiver coils (flexible multipurpose coils or human knee coils) and thus contrast improved the SNR by increasing the number of acquisitions (pulse sequences) per image or improved the resolution by increasing the number of phase encoding steps.

A large selection of coils is available. Flexible coils allow a close contact for good anatomic fit and better resolution. The advantage of making one’s own custom coils is shown in publication V. If an anatomically suitable local coil for a dog that covers only the area of interest cannot be used, a too large coil collects tissue noise also from outside the site of interest. This decreases spatial resolution in the image.

The 0.1T imager also uses a narrower bandwidth than the 1.0T imager that means that the scanning time was longer in 0.1T than in 1.0T but this did not prevent taking all the scans needed for each patient.

**Technical disadvantages**

The resolution is not as good as in high field. The scanning time is longer than in high field but shorter than in ultralow field. Most of the limitations for diagnosis are in the spine. Because of less resolution and the small size of the spinal cord the white and grey matter cannot be seen in transverse planes as they can in high fields. Intramedullary lesions would be missed and the differentiation between extra- and intramedullary lesions would be difficult.

**Diagnostic quality**

**Brain**

In the normal dog brain (N3), the low-field scans (publication IV, Figures 1 a-c) had less resolution but better contrast than the corresponding high-field images, as grey and white matter could be better differentiated. For low field strength imaging the T1 weighted pulse sequences were for sagittal images 2D GRE 300/20 and for transverse imaging IR 3D 370(185)/20. This IR sequence showed better than the corresponding image with 1.0T (SE 600/15) the differentiation of grey and white matter, in accordance with the experiences in MRI of humans by Doyle et al, (1981), and Mills CM et al, (1984). They used IR especially for this differentiation because of its
sensitivity to distinguish between grey and white matter. The lower SNR in low field, as compared to high field, can be enhanced by increasing the voxel volume. Increased matrix size leads to increased spatial resolution but decreased SNR. Increased FOV leads to increased SNR. The relationship between the matrix and the FOV has a definite effect on SNR by determining the size of the individual pixels within the matrix. The pixel area resolution is calculated by dividing the FOV by the matrix size (Kjos et al., 1987). Moreover the imaging time is proportional to matrix size. In our publication V, (Fig. 3, a-c), for T2 weighted SE sequence, we reduced the matrix size from 256x256 to 150x256 reducing the imaging time to 60% of its previous value. As we increased the slice thickness from 3 to 7 mm the SNR increased by a factor of 2.3. In addition, we could further increase the SNR with our new coil.

The 1.0T images had better resolution giving a more pleasing appearance, but did not give more information (publication IV, Fig. 1 a-d).

Spine

The images made of the lumbar spine with 0.1T imager and a standard loop shaped flexible coil placed around the area of interest had better resolution than those made with the ultralow-field imager and the standard spinal coil. There were many more sequences available, and they were much faster than those used in ultralow field. There was also the possibility to make dorsal scans of the spine. The resolution was not good enough for seeing the nerve roots or the internal structure of the cord in transverse scans.

**HIGH FIELD STRENGTH**

**Technical advantages**

The high field-strength imager (1.0T) had an advanced computer design compared with the older ultralow and low field imagers we used. They have ready programmed a larger selection of standard sequences for humans than in low and especially ultralow fields. But also in the high field imaging these sequences do not fit automatically for dogs. For dogs the most probable ones have to be tested and saved if they work well or new ones must be designed. The larger number of sequences allows the best contrast for different lesions. The slice selection plane is totally flexible with minimal thickness of 2-3 mm. Multislice contiguous imaging is possible without any interslice gap or interslice gap can be inserted to reduce overall examination time. The FOV can be selected to give the best result for specific applications. There is a good selection of human coils available to maximise the picture quality in humans but only the knee coil for brain imaging and spinal coil for spinal imaging in dogs were useful. The good resolution gained with these coils and the sequences available give good resolution to see also lesions inside the spinal cord.

**Technical disadvantages**

The major disadvantages are: high cost of installation and maintenance because of helium gas, shielding and weight of equipment, and the disturbing acoustic noise. There is a danger of projectiles and heating of ferromagnetic objects near the magnet. The long narrow bore of the machine makes observation and treatment of the canine patient difficult. The very small volumes of sedative infusions for small dogs cannot be done accurately with long infusion tubes. If an injection is to be made eg sedative or contrast given, the patient has to be taken out of the bore, treated and then repositioned. Then the scout scans need to be repeated. Thus the manner in which immobilization is done is critical for the success of the expensive high field machine. Its short examination time could be negated by the long time required for repositioning the patient and doing new scout examinations. However the latest open bore machines do not have these disadvantages.

For dogs the sequences meant for human use which are likely to work also in dogs have to be tested and saved, or new ones must be designed. This is very time consuming.

All the large number of standard coils cannot be used for dog imaging because of the anatomical differences. No flexible multipurpose or specific coils for canine brain or spine imaging were available. Better results could be obtained with these but the production is very expensive.
**Diagnostic quality**

In our study (publication IV) the high field strength (1.0T) imaging of the same dog as with low field imaging did not give more information for diagnosis in this case or in the case of a brain abscess even when the resolution was better and the acquisition time was shorter. This cannot be generalized. To my knowledge there are no other veterinary publications comparing low field and high field results in the same patient. The majority of the veterinary MRI publications now deal with high field and prove this system to give very satisfactory image quality (for example Kraft et al, 1989,1990,1997; Kornegay 1990; Moore et al 1991; de Haan CE et al,1994; de Haan JJ et al, 1993; Morgan RV et al, 1994, 1996; Kibberger et al, 1997; Kippenes et al, 1999; Levitski et al, 1999 a), 1999b); Mellema et al, (1999). The mid field imaging has also a favourable evaluation of diagnostic quality by Adams et al, (1995), and Lotti et al, (1999). The reports from low field imaging are few but the reports indicated these machines were good enough for confirming a clinical diagnosis. Armstrong et al, (1983) used MRI together with CT to verify the cortical atrophy and enlargement of the ventricles in English setters from 15 months old with canine ceroid lipofuscinois of brain compared to normal.

We had similar finding in P6 with the same disease. We could not see details in white matter like these authors. Hatchcock et al, (1996) found low field scans helpful in evaluation of meningiomas showing common changes like increased signal intensity in T2W images, and decreased signal intensity in T1W images, marked contrast enhancement, and presence of oedema. We did not find meningiomas in our patients.

**Brain**

The brain images with 1.0T showed some details in the margins and small enhancement around the ventricles of a frontoethmoidal meningoencephaloce (P18) better than in the low field images of the same dog. The expanding lesions caused by brain abscess (P4) were not specific enough to provide a diagnosis even with high field imaging. They could be identified only as a mass lesion as Bydder (1984) describes. From our experience with other patients’ high field images we can agree with the authors using high field strength imaging that it has more potential to show small lesions. The small brain infarcts we saw in a dog with borreliosis (P30) might not have been visualized with lower field units. False negative diagnosis in small tumors like meningiomas are likely to occur at ultralow field MRI in humans when 10-15 mm slice thickness is used (Kormano et al, 1987). If this is the case in human brains which are much larger than dog brains we must remember this when evaluating ultralow field images in dogs. The detection of very small lesions in connection with seizures is important if reversible MRI lesions like in human brain should be followed up in dogs. Mellema et al, (1999) found MRI evidence in dogs of reversible lesions in histologically proven lesions like oedema, neovascularization, reactive astrocytosis, and acute necrosis. We had no possibility to follow up imagnings or biopsy in our patients with seizures that had no detectable lesions with our level of technique.

**Spine**

The anatomy of the spinal cord was seen with differentiation of the grey and white matter in transverse section. This was not possible in lower field strength machines. The spinal tumour seen in all three planes in our patient (P45) imaged with high fields could give very exact information of the spread of the tumour. This is in accordance with Adams et al, (1995), who used a mid field scanner, and de Haan JJ et al, (1993) with high field scanner. They state that the transverse T1 images gave the best assessment of the spinal nerve or spinal cord compression, and lateralization of lesions.

In this study we did not evaluate patients with classic type-II disk prolapse so we did not try to show how the lateralization would look like in transverse sections in high field images. Levitsky et al, (1999) a) used highfield scans to show to which side of the spinal canal the disk had protruded. We did not have patients with discospondylitis. The MRI findings of this disease in dogs are described by Kraft et al, (1998). They saw in T1W images a destruction of the vertebral end plates and decreased intensity of the affected
disk. In T2W there was increased intensity of the
disk and end plates.

The chemical shift artifact that Kippenes et al.,
(1999) saw in their patient’s transverse high field
scan did not occur in our patient’s “myelographic”
transverse spinal study with low field imaging.
Chemical shift artifacts increase as the field
strength increases. They are caused by
difference in resonance frequency experienced
by protons in different environments like fat and
water. The signals from fat and water at the same
position result in a relative shift of one of the signal
components (Rinck 1993).

**CONTRAST MEDIUM**

The total examination time was increased in all
field strength machines as an additional T1 scan
was necessary after giving the contrast medium.
This extra time needed and the contrast material
increased the cost of imaging.

**Diagnostic value of contrast medium**

In our patients the application of contrast helped
to enhance a brain astrocytoma in P2 that was
isointense with the white matter and to show the
brain infarcts in borreliosis. Contrast studies were
useful in excluding detectable brain lesions like in
dogs with seizures (P21, 34) or ataxia (P22) that
had negative T1W images and negative contrast
studies. In P31-32 with hydrocephalus seen in
T1W images contrast medium was applied to
exclude lesions in the brain substance that might
have caused the hydrocephalus. We used contrast
medium to see if there were findings with
interrupted BBB. In our patients with
encephalomalacia (P7) and brain abscess (P4)
we could not see enhancement. We could not see
the lesions in P7 most probably because of the
poor resolution of the ultralow image and we had
no experience of earlier similar cases to compare.
The lesion in P4 was possible to see in ultralow
and high field strength scans but we had never
seen such earlier and this was not an experimental
induced abscess. So we were not as successful as
Runge et al., (1985) b) who used Gd-DTPA in
dogs with experimental brain abscesses. They
imaged all with 0.15 T, 0.5 T, and 1.5 T prior to
and following iv bolus injection of 0.25 mmol/kg
Gd-DTPA. The contrast enhancement was
similar at each field strength and the central area
of infection could be discriminated from
surrounding cerebral oedema. Looking at our
scans of P4 we would still keep the possibility of
other mass lesions like tumours in mind.

Drayer et al., (1985; 1989) noted maximal
enhancement immediately in meningiomas and
gliomas and delayed in an infarct model in dogs.
Yoshida et al, (1989) began imaging 5 minutes
after Gd-DTPA injection. They found that the
contrasted MR images should be taken from 5
minutes on as different tumour types have
different recovery times of contrast (different T1
values). T1 time was shortest in meningiomas
where Gd-DTPA leaks easily into surrounding
tissue.

A negative contrast examination cannot
exclude all lesions. Different types of brain lesions
enhance differently in dogs (Kornegay 1990;
Kraft et al, 1990, 1997); Bertoy et al, 1992;
Duesberg et al, 1992) On the other hand very high
levels of contrast can result in loss of signal
intensity (negative enhancement). Contrast
enhancement in MRI is more difficult to interpret
compared to enhancement with CT.

The diagnostic value of the contrast medium
examination is high in veterinary MRI. Drayer et
al., (1985) saw abnormalities in 7 dogs better
detected and characterized after infusion of
contrast. Runge et al, (1985b); Kornegay 1990;
Kraft et al, (1990, 1997), and Stewart et al, (1992)
all reported on benefits of using contrast in their
patients.

**Safety of the contrast medium**

None of our dogs showed any adverse clinical
reactions with use of this contrast medium.
Anaphylactic reactions in humans against
gadolinium contrast agents have been less
frequent than when conventional ionic radiological
contrast media have been used (Niendorf et al,
1991; 1994). The experience is still limited in
veterinary medicine but until now no reports of
complications have been published using similar
doses of contrast as we did. We used contrast in
our studies 0.1-0.2 mmol/kg. This is comparable with other studies. Higher doses lead to loss of enhancement in contrast studies. The contrast agent is also expensive.

Mendonca-Dias et al, (1983) and Slutsky et al, (1985) showed with laboratory dogs that 0.1 mmol/kg Gd-DTPA can be administered safely either rapidly or slowly intravenously without significant haemodynamic sequelaes.

**IMMobilization for the MRI Examination**

Anesthesia with endotracheal intubation using an anesthesia machine was not possible as the imaging had to be performed in examination rooms for human use and no special devices for dogs of non-ferromagnetic material were available. Sedation with medetomidine and 1-methadone proved to be a good combination for immobilization. This failed only in two patients with cerebellar atrophy scanned with ultralow field strength (0.02T) when intensive tremor caused motion. Although imaging times were sometimes 15 minutes long per sequence, our sedation method allowed us to lengthen the immobilization time when needed with repeated im or iv injections. The loud noise of the scanners did not disturb the dogs. The patients had to be transported immediately after the examination back to the animal hospital: a quick recovery with the antidote made this possible. Ketamine was unsuitable as the results from its use elsewhere in the clinical experience in our animal hospital were discouraging and there was no antidote for it. Moore and al., (1991) anesthetized dogs with iv pentobarbital for MRI. Stewart et al, (1992) and de Haan JJ et al, (1993) sedated dogs for MRI with butorphanol tartrate and acepromazine maleate and anesthetized them with pentobarbital sodium. De Haan CE et al, (1994) used general anesthesia induced by sodium thiamylal and maintained with endotracheal intubation with isoflurane in oxygen for imaging healthy laboratory dogs. Levitski et al, (1999) used diazepam and propofol for induction of isoflurane anesthesia. All these mentioned anesthesia methods are widely used and accepted but they need special instruments, patient monitoring device, and a long aftercare in recovery. We could not provide these things in the circumstances we operated. Kraus et al, (1996) advised to use gas general anesthesia. They also mentioned the difficulty of using it in a rented facility and recommend propofol drip or other intravenous protocol with intubation to prevent the side effects of increased carbon dioxide levels and brain pressure. We did not want to use propofol because we did not have possibility for exact monitoring of the respiratory and cardiac systems. Propofol would also been much more expensive to use than the sedatives. It was not even available in the earlier years when the imagings were done.

**Positioning of the Dogs and Coils**

**Positioning for the brain scanning**

Some dogs were imaged with the knee coil in lateral recumbency both in low and high field-strength scanners. Knee coils are either left or right, and are located near the side of the patient couch to accommodate the human form. When brain imaging was made with the knee coil with the dog on its side (lateral recumbency), because of this eccentric location and its high position the loose subcutaneous tissue of the dogs allowed the skull to rotate, making exact straight lateral positioning practically impossible. The best symmetry with all kinds of local coils could be achieved with the dog in the prone position (ventral recumbency). Supine positioning (dorsal recumbency) was not done in any dog as without intubation the breathing would have been compromised. Possibly the rapid sequences of the high field strength could have allowed supine positioning. Also, symmetric images were easier to achieve when the head could be rested on the mandibles. Levitsky et al, (1999a, 1999b) could position the dog patients for cervical spine imaging in dorsal recumbency as inhalation anesthesia was used. The head coil we first tried to use was not good because of its too large diameter for the canine head as its shape was designed for the larger human skull form. Also we could not observe the dog during imaging because the coil’s helmet-like end obscured the head.
Positioning for the spine scanning

The positioning of the dogs for spinal imaging was difficult with all the machines. The same position as for human patients, supine, could not be used as the dog’s breathing would have been compromised without intubation, and prone position could cause imaging artifacts if breathing movements should occur during a sequence. The dogs could be stretched out in lateral recumbency only for cranial thoracic or lumbar part of the spinal imaging. The thoracolumbar curvature of the spine allowed only lengths of 1-2 vertebrae to be included in the same sagittal image. The flexible multipurpose coil used in the prone position in the 0.1 T device was the easiest to use for lumbar spinal imaging.

NEW COIL DESIGN (PUBLICATION V)

From Table 1. in publication V can be seen that the highest SNR improvement obtained with the brain coil compared to the knee coil images occurred at the mid brain area of a dog. This indicates that when the brain coil is positioned on the head of a dog (publication V, Fig.2), its effective centerpoint coincides with the central part of the animal’s brain, which is the position of choice.

If an anatomically suitable local coil for a dog covering only the area of interest is not available, a too large coil collects tissue noise outside the volume to be examined. This decreases spatial resolution. The flexible multipurpose loop-shaped local coils of the 0.1T low field-strength scanner gave the best images of dogs’ head and spine compared with all the ultralow- and low field strength units’ standard coils. This kind of coil was the second best, after the brain coil we designed for Beagle brain imaging (publication V). The use of a dedicated coil resulted in an improved SNR compared with the images obtained with the standard. Since both the custom brain coil and standard coils used for comparison are provided with integrated preamplifiers, the brain coil’s higher SNR is a result of a superior wire geometry. Higher SNR can be used either to improve the final image quality or to shorten imaging time. The latter effect is quadratic; thus a 20 % improvement in SNR (publication V, Table 1) translates into 36 % imaging-time-saving without a compromise in image quality. The time saving becomes significant e.g. if one wants to make an experimental long follow-up study with a large number of research dogs. This coil could also be used for most dogs in a busy MRI facility to save time, or to improve image quality.

OBSERVATIONS ABOUT THE MRI FINDINGS

Anatomical variability of dog brain

The form of the canine brain in MRI for different breeds has some variations depending of the shape of the calvarium. The brachycephalic (and dome-headed) dog’s olfactory brain area is very short and the cerebellum is more under and between the hemispheres than in other types of dogs. The shape predisposes to hydrocephalus (Oliver et al, 1997, Dennis 1998). We found it difficult to determine the normal size of the lateral ventricles as the variations in size and symmetry occurred in a doligocephalic control dog (N3) without any evidence of neurological dysfunction. Other authors have also found that variations in normal dogs in the size (8%) and symmetry (31%) of lateral ventricles occurred in normal laboratory Labrador type dogs (de Haan JJ et al, 1994). Sachie et al, (1997) found a clinically insignificant ventricular enlargement and asymmetry in their Beagle-type laboratory dogs. It is known in human neurology that hydrocephalus may not cause clinical problems if the condition is not caused by an underlying disease or malformation leading to an increased CSF pressure. Vullo et al, (1997) developed a quantitative MRI method to diagnose ventriculomegaly in Beagles. The determination of in vivo ventricular volume was possible in a group of English bulldogs with help of MR images (Vite et al, 1997). This method will help to evaluate the normal ventricular volume as fraction of intracranial volume in different breeds. The measurement of the intracranial pressure would be needed in dogs before making diagnosis of a CSF circulation disturbance as a cause of paraplegia or a progressive brain disease an MRI determined anatomical change in ventricle size does not alone verify it.
The evaluation of the lateral ventricular size requires an exact symmetrical positioning of the dog. An asymmetry of the ventricles can be present without an underlying disease. In our patient P21 this kind of asymmetry was interpreted as a normal variation by the pathologist. The MRI finding of ventricular asymmetry can be caused by a lesion or anatomic variation.

The appearance of borreliosis in brain images

In man multiple focal areas of increased signal intensity in white matter both in proton density and T2 weighted images were seen in borreliosis by Belman et al., (1992). Kaminsky et al., (1998) saw normal MRI in another patient with borreliosis. The focal lesions seen with our patient had similar appearance in T2W images and showed enhancement with contrast in T1W as sign of BBB leakage. In man borreliosis causes vasculitis with secondary thalamic brain infarction (Keil et al., 1997). The MRI differential diagnosis consists of focal or multifocal changes caused by a variety of lesions that cause rupture of BBB and demyelination: infarcts caused by other diseases such as acute disseminating encephalomyelitis, multiple sclerosis, focal infections of other origin, and tumours (Demaerel et al., 1995). Most of these diseases also occur in dogs. Our MRI findings were similar as Belman et al., (1992) describe in children. The other authors that are mentioned here saw variable MRI findings in humans. There are not to my knowledge publications about MRI of borreliosis in dogs. Dennis (1998) says that as brain metastases are rare in small animals multifocal lesions in MRI strongly suggest inflammatory disease. Like we, she uses CSF analysis for confirmation in such cases.

The appearance of brain tumours in MRI

Tumours are generally hypointense in T1W and hyperintense in T2 W images because they often contain more (water) hydrogen protons than the surrounding tissue (Felix et al., 1985). The earlier high expectancy of tumour tissue specificity (Damadian 1971), was later revised to general lack of diagnostic specificity in T1W, T2W and PDW images (Bottomley et al., 1987, Just et al., 1988). Stewart et al., (1992) saw brain tumours in dogs show great variation in signal intensity after contrast application. The appearance of different stages of the same type of tumour can look different in MRI. T2 weighted scans show the changes in tissue character and oedema with better detail than T1 weighted scans even in absence of contrast enhancement (Kraft et al.,1990). One of our astrocytoma patients (P16) showed this feature. In cases of suspected neoplastic mass if the patient’s severe condition allows time only for a single pulse sequence to be used, T2 weighted images are preferable, as exemplified in patient P5 (malignant ependymoma). The signal intensity patterns may be different using high- or low field strength scans (Kraft et al., 1997). Tissue biopsy is needed for definitive diagnosis of intracranial tumours in dogs (Kraft et al., 1997).

Two brain tumors histologically diagnosed as astrocytomas had following appearance in ultralow field strength examinations: in T1W images these tumors had slightly decreased or mixed signal intensity areas (P2 and P3). Post contrast Gd-DTPA images intensely enhanced the tumour of P2. Contrast images gave best determination of tumour size and localization. The third astrocytoma (P16) was seen in high field scans with low signal intensity in the T1W image and high signal intensity around a less intensive inner part in the T2W image (FIG. 13 A-B) similar to the findings of Kraft et al., (1990;1997). Contrast enhancement with Gd-DTPA does not always occur in dogs with astrocytoma if the tumour infiltrates diffusely without surrounding oedema (Kraft et al., 1990; Stewart et al., 1992). Glions can vary in their appearance considerably, based principally on biological behaviour (Kornegay 1990). Glioblastoma multiforme in P45 appeared in our study with 1.0T scanner as hypointense relative to the spinal cord in T1W images (FIG. 22 A-B) Our pathologists did not classify other astrocytomas.

MRI did not show the full extent of diffuse cerebral and leptomeningeal astrocytomas because of absence of oedema and
submacroscopic spread of tumour and lack of contrast enhancement (Kraft et al, 1990).

The distinction of tumour tissue from adjacent peritumoral oedema in MRI is difficult. T2W images obtained with SE pulse sequence and TE over 100ms or application of contrast media Gd-DTPA (Drayer et al, 1985; Felix et al, 1985) can help to this differentiation.

The malignant ependymoma (P5) in the highfield T2W scans was also hyperintense and showed cyst formation as Kraft et al, (1997) described, but was not intraventricular and did not have smooth margins like they saw but was irregular. The ependymomas are ventricular tumours and are seen intraventricularly usually. Possibly our images could not show details enough because of the small size of the dog (9kg) to detect the spread was also in the ventricle. The spread of high signal intensity widely in white matter was most likely oedema around it. The pathologic reports were not sufficiently detailed to verify this.

**MRI evaluation of bone**

The cortical bone had signal void and appeared black in the images. Also new bone formation in spondylosis and sclerotic changes were equally black. The bright high signal from fat could indicate due to its local disappearance a thinning of the cranial bone in the patient with frontal meningocele. Even when MRI is not sensitive in showing bone reactions as well as radiographs it helped us to see that the brain tumours we saw did not invade through the bone like Vogens et al, (1995) describe. We could identify destruction of bone in the patient P45 with glioblastoma multiforme in the spinal canal. A bone like the first cervical vertebra of a dog (P35) with a small intramedullary fat area has not contrast enough to show intraosseous lesions with the low spatial resolution of an ultralow field scanner. This explains the failure to show a whiplash fracture verified in necropsy.

**Cervical spine compression**

Sether et al, (1990) described the MRI characteristics of normal and degenerated disks in dogs. They found, as did this study, that the nucleus pulposus of normal disks appeared with high signal intensity in the T2 weighted images and with midrange signal intensity in T1 weighted images (Fig.1-2,4-5, publication III) owing to the high water content of the nucleus pulposus. In T1W images, annulus fibrosus and adjacent ligaments had as low signal intensity as the cortical bone of the endplates so they could not be distinguished. In disk degeneration, the loss of hydration leads to a reduced signal intensity of the disk in human beings and in dogs (Sether et al, 1990). These authors also used Gd-DTPA contrast for enhancement of the sclerotic endplates of the vertebrae and portions of partly mineralized disks. We did not use contrast medium as we did not think it would improve the diagnosis. In our patients (Publication III, P3-5) MRI proved to be more sensitive in detecting disk degeneration of non-mineralized disks than did radiography, which makes it especially valuable in the diagnosis of fibroid disk degeneration. Even mild herniations of L7-S1 disks could be seen in sagittal MRI scans with all the field strength machines we used.

Even though totally degenerated and desiccated disks and osteophytes extending into the epidural space could not be distinguished because of the similar low signal intensity (black) from each other in T1 weighted sagittal images, the compression of cauda equina could be seen on myelography and in MRI in P47 (=P2 in publication III), where osteophytes and hypertrophied intervertebral ligaments in the epidural space could be identified as an area of diminished signal intensity. The osteophytes, ligaments and degenerated disks have similar low signal intensity and cannot be distinguished from each other in an MR image but if they compress the spinal cord their low signal intensity against the higher signal intensity of epidural fat or spinal cord (or against the high signal intensity of CSF in T2W images) can be seen. Autopsy verified the compressing new bone formation and fibrous hypertrophy. The lumbosacral disk had been removed surgically previously.

We found the MR diagnosis of the cervical spinal compressions due to subluxations of the vertebrae difficult to interpret correctly because the positioning of the neck in such a position that a true sagittal view could show them was not easy.
Whenever the sagittal slice is not exactly in the midline a false positive or negative diagnosis can be made.

Concerning other disorders than wobbler-syndrome caused by subluxation we agree with Sande (1992) that MRI is superior to myelography in the diagnosis of spinal disorders in dogs.

**Spondylosis**

Marked spondylosis seen in radiographs occurred in some dogs (publication III, Patients P10 and 11) without disk degeneration in this area. Morgan J (1967a) however saw spondylosis only in connection with a disk degeneration in his pathologic studies. He concludes (Morgan J. et al, 1967b) that there is a great variety of other conditions that lead to spondylosis formation like degenerative lesions in the joint and in vertebral ligaments. Talvio (1989) found no connection between clinical signs and the site of spondylosis in his canine patients. That implies that either our MRI examinations were not sensitive enough to detect the disk degeneration, or that there are other causes for spondylosis.

**Cauda equina compression**

We used a low field MR scanner in the examination of eleven dogs with lumbar pain and neurological findings and three healthy control dogs before 1992 with success to show cauda equina compressions. Later De Haan JJ et al, (1993) and Dennis (1998) later used high field MRI and Adams et al, (1995) used a mid field scanner, also with success in the diagnosis of cauda equina compression. We agree with them that MRI is more effective and easier to do and interprete than contrast radiographic studies for this purpose.

**THE OPTIMAL FIELD STRENGTH?**

Crooks (1984) says that the optimum field for MRI of humans depends on tissue type, body part, and imaging sequence, so that it has not an absolute value. Hoult et al, (1986) state “We feel that with present techniques, a good argument can be made for a midfield (0.5 to 1.0T) instruments representing a good compromise for composite general imaging of head and torso”. Today it still seems that he was right in his estimation because the development of midfield machines seems to progress both in their technique and popularity. The technique of high field strength machines has been advancing at least even more but their high cost make them not as realistic an alternative for dog studies. In cases where a neurosurgical operation is an option the maximal resolution of neuroanatomy in an image is required. In our investigations the high field strength (1.0T) imager was the optimal one in this matter especially in spinal cord transverse and dorsal section images. An ultralow or low field unit would not be completely satisfactory for this purpose.

The low field strength devices are interesting for veterinary use because of their lower price and maintenance cost. The development has been very rapid also in the low field strength devices and they

"follow by natural progression”. I found it much more difficult. There was very little known in literature about veterinary MRI and even less about low field imaging.

The experience documented in this thesis shows that this is great oversimplification of the requirements for successful MRI examinations of the CNS. Firstly, to ensure immobility and correct positioning for images of diagnostic quality, familiarity with the equipment being used and its special demands are necessary. Then, for correct interpretation, a knowledge of the appearance of various normal and pathologic tissues when using different pulse sequences is a basic requirement, as well as familiarity with artifacts. This knowledge in turn requires an understanding of MRI physics, and an understanding and cooperation between the veterinarian and physicist or technician operating the machine. Thus the natural progression is actually quite a steep learning curve.

**THE INTERPRETATION OF THE MR IMAGES**

Sande (1992) reviewed the available imaging methods for examining the spine and, perhaps in a burst of early enthusiasm for the new method, emphasized in his summary that to be successful, only the variations of normal anatomy needed to be known, and the skills of interpretation would
are getting faster all the time with new technical improvements and more powerful computers. The longer imaging time compared with the high field units seems not to be very disturbing and if an anesthesia machine can be used a long scan time is not a medical obstacle, but may be an economic one – which is balanced against the capital cost and caseload. The units are still being developed with new features, because of their lower cost and simpler installation. But the limitations of resolution affects examination of details even in humans, with larger spines than dogs, for example in the evaluation of foraminal lesions (Rinck 1993). Also transverse imaging of the spinal cord with ultralow (Fagerlund et al, 1989) and low field strength scanners give poor tissue differentiation and inability to angle the imaging plane parallel to a disk even in humans. We had the same experience with dogs.

Owner’s response

The owners of the patients that came in very severe and often advanced state of disease for imaging were pleased with the non-invasiveness and unstressful nature of this procedure. We were fortunate not to lose a single dog or worsen any dog’s clinical condition in our examinations. MRI gave a good support to establish diagnosis, prognosis and therapy plan. This helped the owners to decide about the future of their dogs. So some of these dogs could live on, to the pleasure of their owners. MRI gave some patients a favorable prognosis or a diagnosis leading to succesful therapy.
Conclusions

1. Effect of machine type

The ultralow field strength machines had the lowest, and the high field strength scanner the best resolution of images. The quality of the image of the spinal cord with all the field strengths was sufficient for showing cord compressions and disk degenerations in sagittal planes, but only the high field strength machine gave sufficient resolution of the spine in the transverse plane.

Only the high field strength scans were detailed enough for planning surgery in other than disk prolapse cases.

The imaging time for T1W scanings was principally longest in the ultralow field scanings and shortest in the high field scanings. Adjustments to the parameters used can reduce scan times on the slower machines, at the cost of lower resolution.

2. Immobilization

The immobilization of dogs for imaging with all the used scanners can be done in sedation with medetomidine and methadon iv.

3. Positioning

The positioning of a dog is best in ventral recumbency for all imagings except when using a spinal coil for spinal imaging, when lateral recumbency is best. An exact symmetric positioning is required for avoiding image artifacts and misinterpretation.

Because the cervical spine could not be imaged in flexed position, induced cord compression because of cervical vertebral subluxation could not be demonstrated as in flexed myelographic examinations.

Although both brain and cord both are CNS tissues, the size and shape of the cord creates more problems than brain in MRI. The whole brain could easily be scanned in all directions.

4. Imaging planes

Both imaging directions, transverse and sagittal, are usually needed for evaluation of the brain. An additional dorsal scan can add to identify all the dimensions of a lesion. Imaging only in transverse planes may be sufficient to demonstrate some conditions like hydrocephalus. In all scanning planes, T1W and T2W images should ideally be done. T1W images with contrast enhancement with Gd-DTPA can additionally be made. For MRI of patients in a very poor condition the examination time can be shortened by using only a transverse-oblique T2W scan for diagnosis of possible tumour or oedema.

For spinal cord imaging the sagittal planes show lesions much better than transverse ones in low field strength imaging because of the small size of the spinal cord. For imaging of the whole spinal cord it has to be scanned at least in three different sessions (planes) sagittally because of the curving of spine. If the spinal cord is not positioned exactly straight for sagittal planes there will be artifacts that make the evaluation difficult or impossible. High field strength machine is necessary for planning surgery of tumours. Transverse and dorsal scans in addition to sagittal scans must be made of the spinal canal and cord.

5. Coils

The resolution of an image can be enhanced by using coils in which the area of interest fits best. In all the different field strength MR scanners, the best for brain imaging among the standard coils is the standard knee coil and for spinal imaging a standard spinal coil. With the low field unit, the flexible multipurpose coils or the dedicated brain coil are best for brain imaging. The flexible multipurpose coil is good for low field imaging of the spine.

Our design of the dedicated coil for the dog brain imaging with the low field strength scanner resulted in an improved SNR, as compared with images obtained with the standard coils of the
same 0.1 T imager. This improvement can be used to shorten scan time 30% or improve image quality. It demonstrates the feasibility of designing custom coils for animal patients.

6. Sequences

In all field strength imaging T1W, T2W, and if required, T1W scans with contrast should be made. T1W scans are most useful to show anatomy and lesions of CNS tissue. The T2 weighted images show the disk degenerations (water content of nucleus can be seen bright) and spinal cord compressions (because of the visibility of epidural fat and CSF as myelographic effect) especially well.

The application of MRI requires a good cooperation with hospital physicists and veterinarians to develop sequences which give the highest image quality as possible, regardless of the type of machine. The sequences are selected to serve best the purpose in each dog patient. This work is as necessary investment as the equipment.

MRI is a useful tool for veterinary medicine in the diagnosis of CNS disorders in a noninvasive way.

The specificity of the findings however is influenced by a large variety of factors like: vascularity, necrosis, mineralization, cell behaviour, artifacts, matrix size, slice thickness (partial volume effects), and variations in the selection of voxels by the operators. The interpretation must be based also on clinical examination, the morphology and possibly on follow up images of the same finding as all lesions do not have specific characteristic signal intensity in MRI, and can only be verified with histological examination. Lesions like brain malformations and large ventricles could be easily identified in scans regardless of the weighting or other technical variations in imaging.

The ideal veterinary MR machine should have following features: low cost, easy to install, maintain and use, high resolution, fast imaging sequences, real time imaging, possibility for interventional procedures like biopsy taking, open large short bore (in horizontal and vertical position), large collection of custom made coils for different sizes of dogs, safe to use, low acoustic noise level. It should also be possible to upgrade it later as the technique of MRI advances. None fulfils all of these criteria and because of the cost factor, none is ever likely to.
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