High resolution magnetic resonance imaging anatomy of the orbit

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Chapter 1

INTRODUCTION

"The discovery, refinement, and present sophistication of radiographic imaging has shifted orbital diagnosis toward the modicum of technology and away from almost sole reliance on ophthalmological assessment. In fact, computed tomography and magnetic resonance imaging are the most important advances in orbital diagnosis of the present century."\(^4\)

Imaging techniques

Imaging of the orbit necessitates sophisticated techniques because of its anatomical complexity. Besides standardized echography which is out of the scope of the present thesis, computed tomography (CT) and magnetic resonance imaging (MRI) have become the most important diagnostic tools for the evaluation of orbital disease. Each of these imaging modalities has its advantages and disadvantages. Whereas computed tomography (CT)\(^6\) provides an excellent depiction of the complex bony anatomy of the orbit, MRI enables better resolution and differentiation of soft tissue structures. In contrast to CT, MRI allows for multiplanar imaging without the need of repositioning the patient. CT uses ionizing radiation to produce cross-sectional images of the body whereas MRI is based on the nuclear magnetic resonance (MR) effect.\(^4\) This phenomenon that had first been described in 1945 by F. Bloch and E. Purcell who shared the Nobel prize for physics seven years later, can be explained as follows: nuclei with a net magnetic moment, such as hydrogen ions (protons) which are abundant in living matter, line up parallel in a strong magnetic field and change to a higher energy level when a radio frequency (RF) pulse is applied at right angles to the static magnetic field. The strength of the static magnetic field of clinical MR-scanners ranges between 0.5-2 Tesla. Once the RF pulse is turned off, the nuclei relax to the original energy level and release a RF signal that can be detected using a RF receiver (..coil\(^*\)).\(^3\) This signal is affected by intrinsic and extrinsic parameters:

In proton-MRI, intrinsic parameters include the proton density of the tissue and the tissue-specific constants \(T_1\) (spin-lattice relaxation time) and \(T_2\) (spin-spin relaxation time). The relaxation times that are exponential decay time constants of the nuclear relaxation process, depend on the mobility of the protons in the examined substance. \(T_1\) and \(T_2\) of free water (highly mobile protons) and water-containing fluids (e.g. cerebrospinal liquor, aqueous humor, vitreous body) are high, whereas the relaxation times of tissue containing a greater amount of bound water or less mobile protons (e.g. fatty tissue) are relatively low.

Extrinsic factors include parameters that are set on the MR-scanner, such as time of repetition (TR) and time of echo (TE).\(^4\) TR is the time between RF-pulses and TE the time between excitation by a RF-pulse and the measured signal. Thus, the relative contribution of any of the intrinsic parameters to signal intensity can be varied by choosing specific "pulse sequences" in order to achieve "weighting" of a desired parameter. Tissues with higher \(T_1\) values appear dark (hypointense) on \(T_1\)-weighted (short TR and TE) images and tissues with higher \(T_2\) values appear bright (hyperintense) on \(T_2\)-weighted (long TR and TE) images. Cortical bone (non-mobile protons) and fast-flowing blood (inside arteries and many larger veins) give no signal (..signal void\(^*\)) on MRI (Table 1).

### Table 1. Signal intensities of ocular and orbital tissues on \(T_1\)-weighted (\(T_{1w}\)) images and \(T_2\)-weighted (\(T_{2w}\)) images. H=high, L=low, M=medium, V=signal void.

<table>
<thead>
<tr>
<th>tissue</th>
<th>(T_{1w})</th>
<th>(T_{2w})</th>
</tr>
</thead>
<tbody>
<tr>
<td>cornea, sclera</td>
<td>L</td>
<td>L</td>
</tr>
<tr>
<td>aqueous humor, vitreous</td>
<td>L</td>
<td>H</td>
</tr>
<tr>
<td>normal clear lens</td>
<td>M</td>
<td>L</td>
</tr>
<tr>
<td>uvea</td>
<td>M</td>
<td>M</td>
</tr>
<tr>
<td>extraocular muscles</td>
<td>M</td>
<td>M</td>
</tr>
<tr>
<td>orbital fat</td>
<td>H</td>
<td>M</td>
</tr>
<tr>
<td>connective tissue septa</td>
<td>M</td>
<td>M</td>
</tr>
<tr>
<td>vessels with fast-flowing blood</td>
<td>V</td>
<td>V</td>
</tr>
<tr>
<td>nerves</td>
<td>M</td>
<td>M</td>
</tr>
<tr>
<td>cerebrospinal fluid</td>
<td>L</td>
<td>H</td>
</tr>
<tr>
<td>cortical bone</td>
<td>V</td>
<td>V</td>
</tr>
<tr>
<td>bone marrow</td>
<td>H</td>
<td>M</td>
</tr>
</tbody>
</table>

The signal void of fast-flowing blood can be explained by the fact that protons of flowing blood that had been excited by a RF pulse, have left the imaging slice before their signal can be detected.\(^4\)

The appearance of stagnant blood (e.g. orbital hematoma, orbital venous anomalies, intraocular hemorrhage) on MRI depends on the age of the blood. The MR-signal reflects the biochemical composition of hemoglobin which proceeds from oxyhemoglobin, to deoxyhemoglobin, paramagnetic methemoglobin and finally hemosiderin corresponding to hyperacute, acute, subacute and chronic stages of clot breakdown (Table 2).
Specific gradient echo sequences may be used to specifically depict arteries and larger veins with fast-flowing blood as hyperintense structures ("MR-angiography").

In contrast to iodinated intravenous contrast agents used for CT, the paramagnetic gadolinium-based contrast agents used for MRI do not enhance vessels with fast-flowing blood. According to the signal void of flowing blood, the vessels appear dark. The function of paramagnetic contrast agents is to shorten the T1 time of the tissues through which they are distributed, rendering them brighter on T1-weighted images. On T1-weighted images, enhanced tissues may not be distinguished from orbital fat that also appears bright. Therefore, contrast enhancement should be used in conjunction with special MR techniques designed to suppress the signal of fatty tissue (fat suppression).\(^5\)

The amplitude \((A)\) of the received signal in MR imaging is proportional to the concentration of protons \([H^+]\):

\[
A \sim [H^+] e^{-\frac{T_1}{T_1 + T_2}}
\]

The resolution and quality of MR images depends on the signal-to-noise ratio (SNR):

\[
\text{SNR} = \frac{d \times \text{FOV} \times n}{M},
\]

where \(d\) is the slice thickness, FOV is the field of view (measured area), \(n\) is the number of acquisitions and \(M\) is the matrix size. The theoretical resolution can be estimated if FOV is divided by matrix size.

There are many artifacts (e.g. motion, partial volume, chemical shift or metal artifacts)\(^6\) that may deteriorate the quality of MR images. It is important, to recognize these artifacts in order to avoid misinterpretation of the images. Artifacts are briefly described in chapter 8.

In order to receive the MR signal, volume coils (e.g. standard 28cm-diameter head coils) that are placed around the whole head or small-diameter surface coils that are placed directly over the region of interest, are used. Surface coils allow high-resolution imaging of the orbit by increasing the signal-to-noise ratio.\(^7\) The depth from which signals are received by the coil, is proportional to its diameter. The signal drops off with increasing distance of the area of interest from the coil. Therefore, imaging of the orbital apex requires either a larger surface coil, a standard head coil, or both for optimal imaging. When additional imaging of the middle cranial fossa is required, the use of a head coil is recommended.

The signal emitted from different tissues can be localized by gradient coils and processed by computers to produce cross-sections through the body.\(^4\) Additionally, magnetic resonance spectroscopy (MRS) can be performed following MR imaging during the same session. MRS is a unique method of investigation for the visual system because it yields biochemical information on the tissue in vivo.\(^4\)

In the present thesis, spin-echo (SE) pulse sequences were used to produce T1-weighted (TE=14-18 ms, TR=440-620 ms) and T2-weighted (TE=110-120 ms, TR=2500-2800 ms) images on 1 and 1.5 Tesla scanners. The slice thickness was 1-3 mm and slices were orientated in the axial, coronal and oblique-sagittal planes (parallel to the optic nerve). The field of view (FOV) ranged between 140x140 mm and 230x230 mm with a matrix size ranging from 256 x 256 to 512 x 512 pixel. Paramagnetic contrast agents were only applied for intracranial imaging. The emitted RF signals were detected using head coils and monocular and binocular surface coils.

### Aims and Outline of Thesis

The clinical applications of MRI have advanced rapidly over the past several years and many articles on the diagnosis of orbital lesions using MRI have been published. Although MRI has the potential of depicting tiny anatomical structures, detailed descriptions of anatomical structures in orbital magnetic resonance images are provided in very few publications\(^5\). However, a profound understanding of orbital anatomy is a prerequisite for the interpretation of clinical findings on MR images. Additionally, a detailed knowledge of the intricate anatomic relationships within the orbit is crucial for successful surgical intervention in this region.

Chapters 2-5 of this thesis are aimed at describing the anatomy of the orbit on high-resolution MR images. Own results on high-resolution MRI in normal subjects are presented and compared with the literature. Finally, clinical implications of our findings are discussed.

Chapter 6 and 7 deal with the application of MRI to functional-anatomical problems related to the mechanics of the upper eyelid. The eighth chapter provides a correlation of orbital MR images with anatomical cryosections.

The following questions had to be answered by this thesis: (1) Is high-resolution MRI capable of depicting orbital blood vessels and nerves and is it possible to visualize connective tissue septa of the orbit? In this regard, the present thesis represents a continuation of the anatomical work of Koornneef\(^6\) using modern imaging techniques in vivo. These questions are addressed in chapters 2-4 and 8.

(2) The second question arose during anatomical and surgical dissections of the upper eyelid conducted by Priglinger and coworkers\(^5\) who found that Whitnall's ligament\(^6\) actually forms a sling around the levator palpebrae superioris (LPS).
muscle (see chapter 5). The band-like fascia[13] between the superior rectus muscle and the LPS has therefore been called "lower part of Whitnall's ligament"[14]. During ptosis operations, it has observed that the amount of levator muscle resection can be smaller when Whitnall's ligament is not severed." Based on this experience, Anderson and Dixon[15] and later Goldberg and coworkers[16] who performed MRI studies of the upper eyelid, have suggested that Whitnall's ligament would act as a pulley or suspensory ligament of the levator muscle. Moreover, fibromuscular pulleys have recently been described in connection with the recti muscles which course in a curved path through the orbit (see chapter 3).[17][18] Therefore, our second question was, whether the levator muscle also courses in a curved path and wheter Whitnall's ligament may be responsible for this course or in other words whether Whitnall's ligament represents the pulley of the levator muscle. These issues are adressed in chapter 5 and 7.

(3) Ptosis surgeons know that the amount of levator resection always exceeds the achieved amount of lifting of the upper eyelid[12] This lead us to the third question: How is the relation between the amount of contraction of the LPS and the upper lid elevation? The possible causes of this relation that has implications on the dose-response relationship in ptosis surgery, are investigated in chapter 6.

References
