

BRAIN IMAGING at 3T and CHALLENGES at 7

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Introduction:

Magnetic resonance imaging (MRI) studies of the brain can be classified into two general categories based on the information type that is being collected, namely those that seek to obtain morphological information and those that aim to describe physiological and functional properties. In the morphological category, we include imaging of tissue and cerebrospinal fluid (CSF) spaces, vasculature, and fiber bundles (connectivity). In the other category are mapping of brain function (fMRI), perfusion imaging, and diffusion imaging. These are all topics that are separately treated in individual lectures in this course. In this lecture, we will focus on the significance of magnetic field magnitude for these different types of brain imaging efforts.

Morphological imaging of brain tissue:

Anatomical images of brain tissue rely largely on proton density, T_1 , and T_2 differences between different regions (e.g. cortex vs. subcortical nuclei) and tissue type (white matter, gray matter and CSF). The region specific values for these parameters have been reported in numerous papers (1-13) for 1.5 T field strength. Proton density is clearly a magnetic field independent parameter. However, relaxation times T_1 and T_2 are field dependent, generally increasing (14,15) and decreasing (16-18), respectively, with higher magnetic fields (also see review (19)). The conventional wisdom has been that the T_1 not only increases with higher magnetic fields but it actually converges so that the distribution of T_1 values among different tissue types would tend to become narrower; this, of course would predict a lower contrast in the brain at higher magnetic fields. However, this was shown not to be the case starting with the early human brain images obtained at 4 Tesla (20) and later on extended to 7 Tesla (21). In fact, contrary to expectations, the distribution of T_1 values among different tissue types in the brain, including the difference between cortical gray matter and adjacent superficial white matter increase with increasing magnetic field (unpublished results).

Lengthening of T_1 with increasing magnetic field also holds true for blood. Blood T_1 is insensitive to its oxygenation state, and varies linearly with field strength going from 1.5 T to 9.4T according to $T_1 = 1.226 + 0.134B_0$. This imparts a clear benefit in time-of-flight type angiographic imaging, as well as perfusion imaging using spin labeling techniques.

While T_2 value in brain tissue decreases with increasing magnetic fields, the decrease is far from linear. For example, going from 1.5T to 7 T, the cortical gray matter T_2 is reduced by \sim factor 1.6 (e.g. ~ 87 ms (7) vs. 55 ms (18)) while the magnetic field increases 4.7 fold. Interestingly, the apparent T_2 values that are measured in a Carr-

Purcell type sequence with multiple 180° pulses are larger and similar to those observed at the lower fields, suggesting that exchange and/or diffusion in the presence of gradients may be a dominant mechanism responsible for the shortening of T_2 with increasing magnetic field magnitude (16,17). For blood, however, the T_2 decrease is dramatic and goes as the square of the magnetic field ((22) and references therein). While blood T_2 is longer than tissue T_2 at 1.5 Tesla, it is significantly shorter than tissue T_2 at 4 and in particular 7 Tesla.

With the recent interest in and availability of high magnetic field systems, another contrast mechanism that may become more frequently utilized is T_2^* or susceptibility variations among different brain components (e.g. (23,24)). While this mechanism is clearly expected to be useful in visualizing deoxyhemoglobin containing vasculature, i.e. the venous system (24), it actually can provide contrast among brain tissues such as gray and white matter as well (23).

As in all MR applications, image quality, measurement time, and/or spatial resolution for morphological brain images depends on SNR. SNR increases with increasing field strength. SNR, however, becomes rather complex when high magnetic fields (hence high frequencies) are considered with lossy biological samples such as the human head. The relationship between SNR and field strength has been examined for biological samples in numerous theoretical studies (25-31), predicting increases with field strength. Field dependence of SNR was experimentally examined in the human head, initially comparing 0.5, 1.5 and 4 Tesla (~21, 64, and 170 MHz, respectively), using a surface coil, documenting that SNR for the ^1H nucleus increased at least linearly at the higher frequencies (32). More recently, B_1 field profile and SNR was examined in the human head for 4 and 7 Tesla when using a TEM “volume” head coil (21). Using virtually fully relaxed images, the SNR was shown to scale ~2 fold going from 4 to 7T, more than linearly with field magnitude, in the center of the brain and less than linearly in the periphery. The diminished SNR gains in the periphery can be recovered using multi channel array coils (33). Clearly, however, at high fields such as 3T and above, SNR must be considered as a function of location within the head and specific coil geometries. This is because the human head/RF interactions approach “far field” conditions at magnetic fields like 7 Tesla where the wavelengths are comparable to or smaller than object dimensions, and the B_1 exhibits a traveling wave behavior (34-38).

Physiological and Functional Imaging in the Brain:

Perfusion Imaging: Images of perfusion or perfusion changes associated with increased neuronal activity can be obtained using ASL techniques that utilize the water protons in the blood as an endogenous “transient” tag. These methods rely on either continuous (e.g. (39-41)) (or dynamic (i.e. modulated) versions of continuous (42,43)) or pulsed (e.g. (44-46)) tagging approaches. All of these techniques benefit from increased T_1 encountered in higher magnetic fields. This is expected to ameliorate errors introduced by transit delays, extend coverage over the brain and yield higher CNR, and specificity to tissue. Excellent perfusion images based on continuous arterial spin tagging

have already been accomplished at 3T (47) and this approach has also been used for functional mapping at 3T (48).

The tissue specificity of ASL improves because of the fact that tagged spins require a finite amount time to reach the capillaries and exchange with tissue water. At shorter periods, larger blood vessels in the arterial side can dominate the measurement, confounding quantitation (e.g.(49)), and appearing as “activated” in perfusion based functional images. In ASL measurements, generally the tag that can be detected in the veins subsequent to the tag’s passage through the capillaries is thought to be negligible and ignored, even though there has not been an experimental confirmation of this. At high fields, this potential contribution should vanish due to the short T₂ of venous blood.

Diffusion Weighted Imaging: This type of imaging is used either for tractography or clinically for determining the alterations with the overall diffusion properties of water subsequent to a pathological change such as stroke. This approach benefits from SNR gains of high field and but suffers from the shortened T₂ with increasing magnetic field. Nevertheless, since the decrease in T₂ is not strongly field dependent and SNR increases at least linearly, there are gains with increasing magnetic field, provided such gains are not lost in less than perfect hardware performance as the demands increase at the higher magnetic fields.

Imaging of Neuronal Activity (functional brain imaging): In the armamentarium of techniques used for investigating brain function, functional magnetic resonance imaging (fMRI) has come to play a dominant role in both human and animal model studies. Today, functional images in the brain can be obtained using the BOLD mechanism using gradient echoes (GE-BOLD) (50-52), cerebral blood flow (CBF) changes using arterial spin labeling (ASL) (e.g. (45,53-57) and references therein) or intravoxel incoherent motion (IVIM) (58,59), and cerebral blood volume (CBV) changes (e.g. (60-62)). The most commonly used fMRI approach is BOLD mechanism. Magnetic field magnitude plays a significant role with respect to contrast, contrast-to-noise ratio (CNR), spatial resolution, and specificity (i.e. accuracy) of functional images obtained with the GE-BOLD mechanism (see reviews (63-65)). While the SNR gains with increasing field magnitude are significant in being able to obtain high resolution functional maps, the most dramatic impact of magnetic fields is on the accuracy of the functional images and CNR associated with signals that possess the highest degree of specificity with respect to boundaries of neuronal activity. This impact is a consequence of the role played by vasculature of different sizes in mediating MR detectable functional imaging signals.

The deoxyhemoglobin (dHb) changes that ultimately give rise to BOLD effect appears first within the capillaries in parenchyma where neuronal activity is modulated; however, these changes are not static in space and propagate to draining veins distant from the site of neuronal activity. These draining veins contribute to GE-BOLD fMRI at all magnetic fields although the specific mechanisms that dictate their contribution differ with magnetic field strength. However, as the field magnitude increases, the small contribution from the capillaries, representing a significantly more accurate depiction of altered neuronal activity, increases virtually quadratically with magnetic field to attain magnitudes that are detectable now in GE-BOLD fMRI maps. While the GE-BOLD fMRI contains both these non-specific large vessel and specific capillary contributions at

fields such as 7 Tesla, the large vessel effects can be suppressed by using Hahn spin-echoes (HSE) instead of gradient echoes (18,22,66); however, the Hahn spin-echo approach does not work at lower fields due to the contribution of intravascular blood signals to functional maps (e.g. (22,67)).

Field dependence of BOLD based functional images must also take into account the contribution of physiologically induced fluctuations in consecutively acquired images in an fMRI series (68-70). In GE-BOLD images, there is a signal dependent contribution to these fluctuations that increase in magnitude with magnetic field. When these fluctuations dominate over thermal noise the CNR of the functional images, then CNR gains with magnetic fields are ultimately limited. However, for high resolution and high contrast imaging where the thermal noise dominates gains in CNR with higher fields are expected. Furthermore, HSE fMRI does not contain these fluctuations that are proportional to signal magnitude and hence improves in CNR with magnetic field (71).

In contrast to the BOLD based functional images, field dependence is much more straight forward for CBF and CBV based mapping. They come through only SNR gains, and the impact of the T_1 's on the method. For example, for ASL techniques, the long blood T_1 at high fields is an advantage (49) while in the CBV based VASO method (62) the converging blood and tissue T_1 's at the high magnetic fields is a disadvantage.

Parallel Imaging:

Parallel imaging with multichannel receiver coil arrays has rapidly become an integral part of MR imaging in the head and body. Parallel and high-field MRI are particularly promising when combined with one another. This is because the two approaches exhibit a high level of complementarity with respect to their favorable and less favorable characteristics regarding SNR, increased magnetic field inhomogeneities, and SAR. For neuroimaging, one of the most significant field dependent properties of parallel imaging is that its performance in terms of reduction factors that can be achieved increases with high magnetic fields (Wiesinger, 2004 #1927} and references therein). It is therefore expected that for all the neuro applications listed above, parallel imaging will play an important role especially as the field magnitude increases.

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