

Ultrahigh field magnetic resonance imaging and spectroscopy

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1. Introduction

In our laboratory, the effort to pursue high magnetic fields has been intricately tied to our interest in developing methods for the acquisition of physiological and biochemical information noninvasively using the nuclear spins of the water molecules and metabolites in the human body. In this effort, a relatively recent and unique accomplishment has been the introduction of the ability to map human brain function noninvasively. The concurrent and independent work performed at the University of Minnesota, Center for Magnetic Resonance Research [1] and at MGH [2], was, in our case, conducted at 4 Tesla. It was one of the first experiments performed at 4 Tesla in our laboratory. Today, functional images with subcentimeter resolution of the entire human brain can be generated in single subjects and in data acquisition times of several minutes using 1.5 Tesla MRI scanners that are often employed in hospitals for clinical diagnosis. However, there have been advantages in using significantly higher magnetic fields such as 4 Tesla, and recently 7 Tesla in humans, and 9.4 Tesla in animal models. Similarly, over the last two-and-a-half decades, spectroscopy studies in intact cells have proven to be rich in biochemical information. However, the most useful of these studies were performed in isolated cells or perfused tissues. Only recently, they were extended to small animal models using high magnetic fields. In human applications, spectroscopy efforts pursued at the commonly available magnetic field of 1.5 Tesla were in general unable to produce data comparable in information to the high field perfused organ or animal model data. This has changed with the availability of ultrahigh magnetic fields for human applications. While the use of very high magnetic fields such as 7 Tesla in human studies is still in its infancy, the data

gathered to date suggest that there are significant gains for spectroscopy studies in general and some of these accomplishments relevant to high magnetic fields are reviewed in article.

2. Imaging using the hydrogen nucleus

2.1. Signal-to-noise ratio (SNR)

In all NMR experiments, especially in *in vivo* applications, gains in SNR are the key to extending the applications of this phenomenon to new frontiers in research. SNR gains can be achieved in going to higher fields. SNR, however, becomes rather complex when high magnetic fields (hence high frequencies) are considered with lossy biologic samples such as the human body and the human head. The relationship between SNR and resonance frequency, ω , or equivalently field strength has been examined for biologic samples in numerous studies [3–9], predicting increases with field strength. At high frequencies such as 170 MHz (^1H frequency for 4 Tesla) and above, SNR must be considered as a function of location within a sample and for particular sample geometries.

Field dependence of SNR was experimentally examined by our group in the human head, initially comparing 0.5, 1.5, 4 Tesla (~ 21 , 64, 170 MHz, respectively), using a surface coil, documenting that SNR for the ^1H nucleus increased at least linearly at the higher frequencies [10]. More recently, with the availability of a 7 Tesla human system in our laboratory, we examined and compared the B_1 field profile and SNR in the human head for 4 and 7 Tesla when using a TEM “volume” head coil [11]. It was expected that a coil with a uniform B_1 field profile when empty will nonetheless lead to a highly nonuniform B_1 , SNR, and power deposition over the human head and brain. The B_1 distribution was mapped over the human head using mag-

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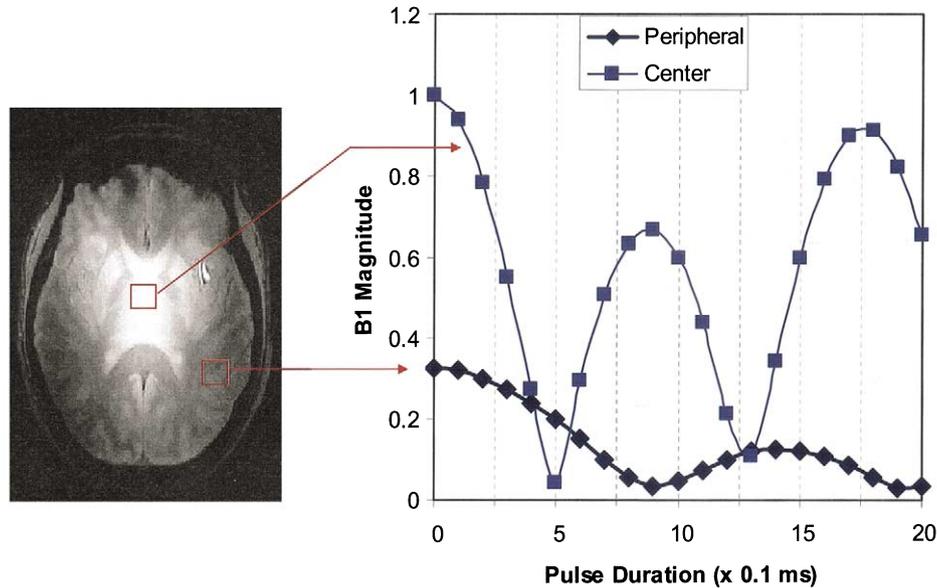


Fig. 1. A, Images of a transverse slice through the human head at 7 Tesla, obtained with magnetization preparation to encode B_1 magnitude as $\{\cos(\tau\gamma B_1)\}$ “turboFlash” (center-out k-space sampling, TR/TE = 4.2 ms/2.5 ms, slice thickness = 5 mm, flip angle = 10° at the slice center, matrix size = 128×64). The different images correspond to different τ ; B_1 magnitude was kept constant. B, The signal (magnitude) oscillation frequency plotted for two regions of interest in the middle and the periphery of the brain, from the data shown in Fig. 1A. Oscillation frequency is equal to γB_1 indicating that the B_1 is substantially higher in the center of the brain. From reference [11].

netization preparation followed by ultrafast imaging (“turboFlash”). Magnetization preparation was accomplished with a variable-duration hard (square) pulse followed by rapid gradient spoiling to eliminate transverse magnetization. The resultant longitudinal magnetization is directly proportional to $\{\cos(\tau\gamma B_1)\}$ where τ and γ are the pulse duration and gyromagnetic ratio, respectively. The results demonstrated that the signal intensity oscillates with higher frequency in the brain center than in the periphery at both field strengths but much more so at 7 Tesla (Fig. 1). At 4 T, the B_1 strength in the brain periphery was down 23% from the center value. At 7 T, the peripheral B_1 was 42% lower than the central B_1 . The nonuniform B_1 profiles obtained experimentally were both expected and predicted by the Maxwell models of the human head loaded TEM coil [11].

Using virtually fully relaxed images when the center of the brain was set to experience a 90° pulse (thus, by necessity the periphery undergoes a less than 90° rotation), the SNR scaled more than linearly with field magnitude in the center of the brain, and less than linearly in the periphery. When averaged over the entire slice, SNR at 7 T increased over 4 T by 1.6 fold without considering a difference in the gain of the 7 T and 4 T coils, and 1.76 times when corrected for the gain difference. Thus, the average SNR for the slice increased linearly. However, the increase was not spatially uniform.

The SNR increases measured in going from 4 to 7 T, were for conditions of same acquisition bandwidth and full relaxation. Such acquisition conditions may be quite appropriate for some imaging applications. If signal averaging is necessary with repetitions that do not permit full relaxation, then, SNR gains at the higher magnetic field will diminish

due to the longer T_1 for some nuclei, such as protons, at the higher magnetic field. Similarly, under many conditions the bandwidth needs to be higher at the higher field strength. For example, in echo planar imaging (EPI), in order to overcome the deleterious effects of increasing B_0 inhomogeneities (and consequently shorter T_2^*), the k-space coverage must be faster. This may be accomplished by scaling the bandwidth with the field magnitude as well, in which case the SNR gain is reduced to square root of the magnetic field ratio. However, solutions to this problem exists in some cases, for example, using restricted field-of-view (FOV) when possible [12–14] or using parallel imaging to reduce the phase encoding steps and thus cutting k-space coverage time.

2.2. Anatomic imaging

Prior to the introduction of 4 Tesla systems in academic laboratories, the expectation was that the contrast in anatomic images would decrease due to converging T_1 values, and would be dominated by the inhomogeneities in signal intensity due to nonuniform B_1 . While the T_1 values have gotten longer indeed [15,16], they have not converged. In addition, the signal intensity variations expected due to B_1 inhomogeneities have been suppressed either by insensitivity to B_1 in the acquisition [17,18] or by postprocessing strategies [11]. Consequently, excellent anatomic imaging has been feasible at magnetic fields of 4 to 8 Tesla in the human brain (e.g., [11,17–25]). Fig. 2 displays an example of a T_1 -weighted image acquired with $0.8 \times 0.8 \times 1.25$ mm³ spatial resolution and total acquisition time of 5 min at 7 Tesla, using a volume coil for transmit and receive.

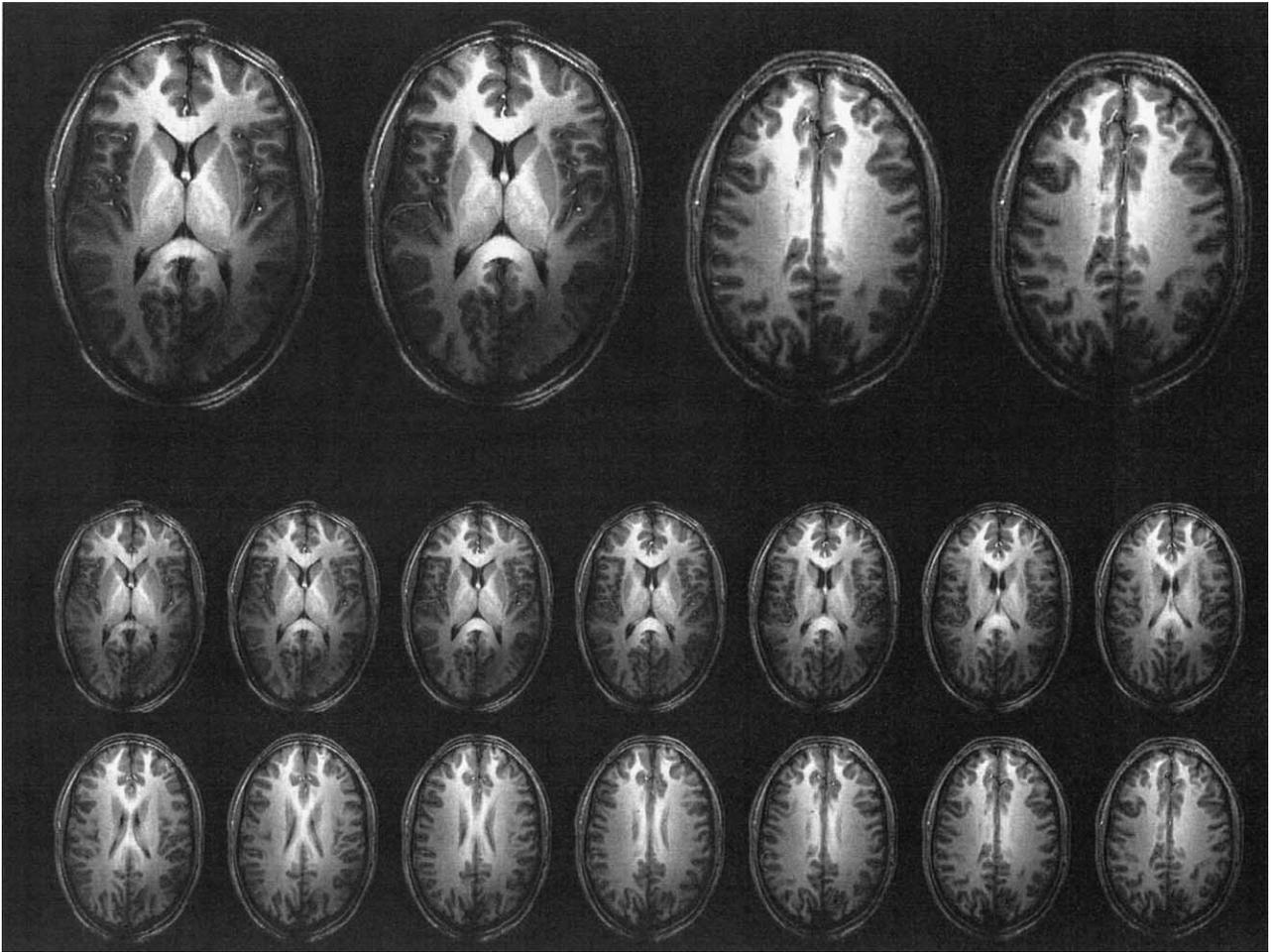


Fig. 2. High resolution ^1H image of human brain ($0.8 \times 0.8 \times 1.25 \text{ mm}^3$) obtained with 3-D acquisition in 5 min. Supplied by Haiying Liu, University of Minnesota.

Improvements on this are expected by the use of multiple coil arrays.

2.3. Functional 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. The most commonly used fMRI approach was introduced in 1992 [1,2,26] and is based on imaging regional deoxyhemoglobin alterations that accompany changes in neuronal activity. This contrast mechanism, referred to as blood oxygen level-dependent (BOLD) contrast, originates from the intravoxel magnetic field inhomogeneity induced by paramagnetic deoxyhemoglobin sequestered in red blood cells, which in turn are compartmentalized within the blood vessels. Magnetic susceptibility differences between the deoxyhemoglobin-containing compartments versus the surrounding space devoid of this strongly paramagnetic molecule generate magnetic field gradients across and near the boundaries of these compartments. Therefore, signal intensities in magnetic resonance

images sensitized to BOLD contrast are altered if the regional deoxyhemoglobin content is perturbed. This occurs in the brain because of the spatially specific metabolic and hemodynamic response to enhanced neuronal activity; it has been suggested that regional blood flow (CBF) increases while oxygen consumption rate (CMR_{O_2}) in the same area is not elevated commensurably [27], resulting in decreased extraction fraction and lower deoxyhemoglobin content per unit volume of brain tissue. Consequently, signal intensity in a BOLD sensitive image increases in regions of the brain engaged by a “task” relative to a resting, basal state. In this approach, the vast majority of images are acquired using T_2^* weighting using gradient echoes. Few studies, especially recently with the introduction of high magnetic fields, have used Han-spin echoes.

High magnetic fields provide the capability for obtaining robust, BOLD-based functional maps with increased contrast, resolution, and specificity (i.e., accuracy). It has already been demonstrated (e.g., see [12,28–31]) that robust, reproducible high resolution T_2^* BOLD maps can be obtained in the human brain at 7 Tesla, and in rodent brain up to 11.7 Tesla [32]. Examples of human brain functional

imaging at 7 Tesla are illustrated in Fig. 3 for two slices obtained with $1 \times 1 \times 3$ mm resolution in the visual cortex. Both negative and positive BOLD changes are clearly observed in these 7 Tesla functional maps (Fig. 3A). The figure also demonstrates the reproducibility of these images in consecutively acquired but separate maps. It can be seen that with the exception of a few pixels at the edges, the functional maps are highly reproducible, especially for the larger positive changes. In fact, the correlation coefficient that determines their significance of activation is also highly reproducible (Fig. 3C). In a separate study, we have reported that $\Delta 1/T_2^*$ induced by neuronal activation was 2.1 fold larger at 7 Tesla compared to 4 Tesla, which led to images with larger areas of activation for the same threshold of statistical significance [30].

High resolution images, however, do not necessarily yield high accuracy maps. Basic image resolution dictated by contrast-to-noise ratio in fMRI (i.e., the magnitude of the signal changes induced by activation compared to image-to-image signal fluctuations), and accuracy in functional images are two independent parameters dictated by different processes. In discussing the spatial specificity of fMRI, we must also distinguish *single condition vs. differential mapping*. Differential mapping refers to functional images obtained by using two analogous but orthogonal activation states designed specifically to eliminate or suppress common signals; the functional image generated in this fashion correspond to a subtraction of the two conditions. In contrast, single condition mapping does not rely on a subtraction from a second orthogonal activated state. Mapping ocular dominance columns by stimulating one eye *vs.* the other in an alternating fashion is differential mapping. Stimulating one eye and using a dark state as the control would correspond to *single condition mapping*. Even if the imaging signals are *not* specific enough to be confined to the territory of one column, a functional map of the columns can be obtained in the differential image if a given group of columns respond slightly differently to the stimulation of each eye and, false (i.e., nonspecific) activation is induced equally by stimulating the two different eyes. Phase encoded mapping of multiple areas using an activation paradigm that cycles through all possible stimulations (as employed in the visual system (e.g., [33–35]) is also a kind of differential mapping because it suppresses common, non-modulating components.

Single condition mapping at a high enough specificity to obtain columnar maps appears unlikely by T_2^* BOLD fMRI at any field strength. It was rigorously demonstrated that T_2^* BOLD fMRI method fails to generate *single-condition* functional images of iso-orientation domains in the cat cortex [36,37] at 4.7 Tesla. Although the image resolution was sufficiently good to detect columnar organizations associated with orientation preference in these studies, they were *not* observed. Furthermore, the images were not complementary when orthogonal orientations were used, and the highest “activity” was associated with a large draining vein,

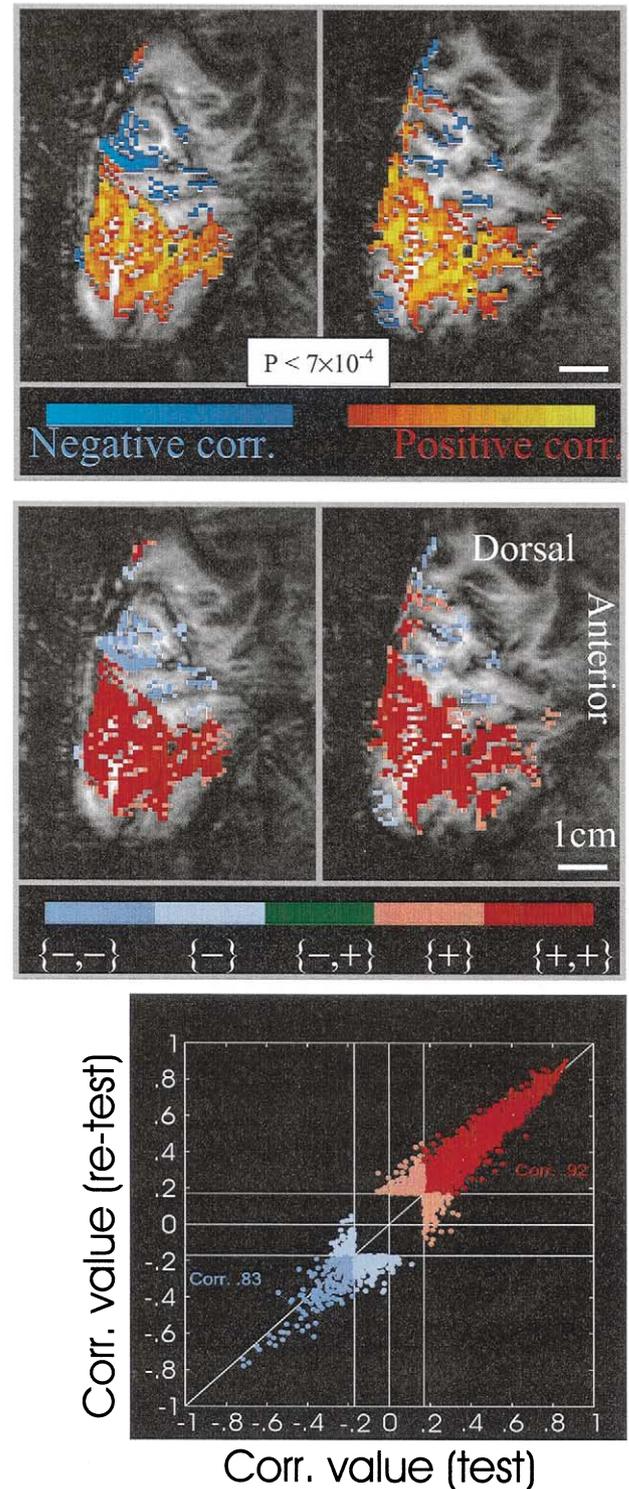


Fig. 3. 7 Tesla, $1 \times 1 \times 3$ mm³ fMRI functional images in two sagittal planes in the human visual cortex obtained with T_2^* BOLD contrast, showing positive and negative signal intensity changes (A). Acquisition parameters were 4 segment gradient recalled echo EPI, TR = 150 ms per segment, TE = 20 ms. Reproducibility of the images are illustrated in (B) for two consecutive runs in the same subject; the pixels that are deemed to have shown a statistically significant increase are color coded as darker red for positive BOLD signals and darker blue for the negative BOLD signals. C, Cross-correlation of all pixels with the activation template for the two runs against each other. Adapted from [28].

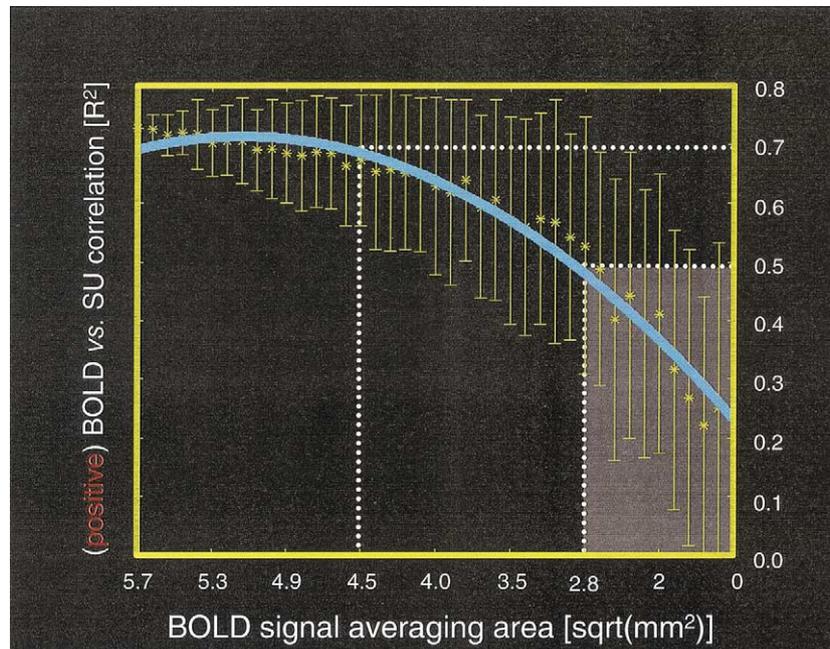


Fig. 4. The effect of voxel size on the relationship between BOLD fMRI signal amplitude and neuronal spiking. The spiking data were recorded from multiple sites over cat visual area 18 covering the activated area depicted in Fig. 1. When all spike data and BOLD data over a the large area of activation were averaged, a linear correlation was seen between BOLD amplitude at 4.7 Tesla and the spiking rate when the stimulation was modulated. BOLD single-unit data pairs from individual recording sites were reshuffled using methods of Monte Carlo permutation. Correlations between n randomly shuffled single-unit sites and their corresponding n BOLD sites (i.e. R^2 between BOLD and single unit responses), were calculated. The mean and standard deviation for an exhaustively large population of random shufflings were calculated at each value of n up to the maximum number of sites recorded, plotted against the nominal voxel size, calculated as $n \times 900 \text{ um}^2$ (cortical area within one voxel). The plot was then fitted with a conventional sigmoidal function, $y = 1/(1.41 + 2.70 \exp(-0.151*x))$. For each voxel size, the distribution of the neuronal qualities is also indicated by the standard deviation. The results depicted in this figure predict that the neuronal correspondence saturates around $R^2 = 0.7$ at the voxel size of around $4.7 \times 4.7 \text{ mm}^2$. Larger voxel sizes are suggested to be ineffective in further improving the level of neuronal correspondence. That is, the maximum amount of variance in the underlying neuronal modulation that can be explained with the variance of conventional T_2^* -based positive BOLD is about 70%. Once the voxel size has been reduced to be smaller than $\sim 2.8 \times 2.8 \text{ mm}^2$, only less only than 50% of the variance in the underlying neuronal modulation can be explained through the observed BOLD responses.

the sagittal sinus. Thus, spatial specificity in the submillimeter domain of iso-orientation columns does not exist in T_2^* BOLD fMRI. Additional limitations on the accuracy of T_2^* BOLD fMRI was provided by multiple site single unit recordings and fMRI studies on the same animal. These studies suggested that the limit of spatial specificity of T_2^* BOLD may be in the 4 to 5 mm range for *single-condition maps* (Fig. 4) [38,39]. In this study, high-resolution T_2^* BOLD fMRI was performed together with single-unit recording at multiple sites. The BOLD response was found to be proportional to the spiking rate when data from all recording sites was averaged and compared with the BOLD response from the entire area where it was deemed statistically significant. However, this correlation started breaking down when smaller and smaller areas were considered for the averaging of the two distinct measurements, in particular when the area started decreasing below approximately $4 \times 4 \text{ mm}^2$. This result is also in excellent agreement with the numbers provided for the accuracy of fMRI maps at 1.5 T human brain using a phase-encoding method [35] where the full width at half maximum was estimated to be 3.5 mm. Consistent with the observations, studies which have reported successful differential images of ocular dominance [41–44] have also reported that single condition mapping

by T_2^* BOLD fMRI was not feasible. Contrary, to all of these findings and the well documented presence of non-specific contributions to T_2^* BOLD fMRI, single condition imaging of human ocular dominance have been reported in one study [40].

The lack of spatial specificity in T_2^* BOLD fMRI might reflect imprecise spatial-coupling between neuronal activity and the physiological and metabolic events that ultimately yield the functional images. Thus, blood flow increases that accompany enhanced neuronal activity might exceed the boundaries of active neurons. Alternatively, or in addition, the lack of spatial specificity could originate from the fact that vasculature plays a crucial role in the generation of MR detectable signals from deoxyhemoglobin changes that accompany neuronal activity. In fact, it is well documented that large “draining” vessels do contribute to the T_2^* based BOLD signals [45].

The accuracy of spatial coupling between neuronal activity and increased blood flow can be evaluated using perfusion imaging (e.g. [46]). This of course requires that perfusion based functional maps do not contain contributions from large vessels. When blood flow to a volume of tissue increases, flow in the large arterial and venous blood vessel that supply or drain blood from that particular vol-

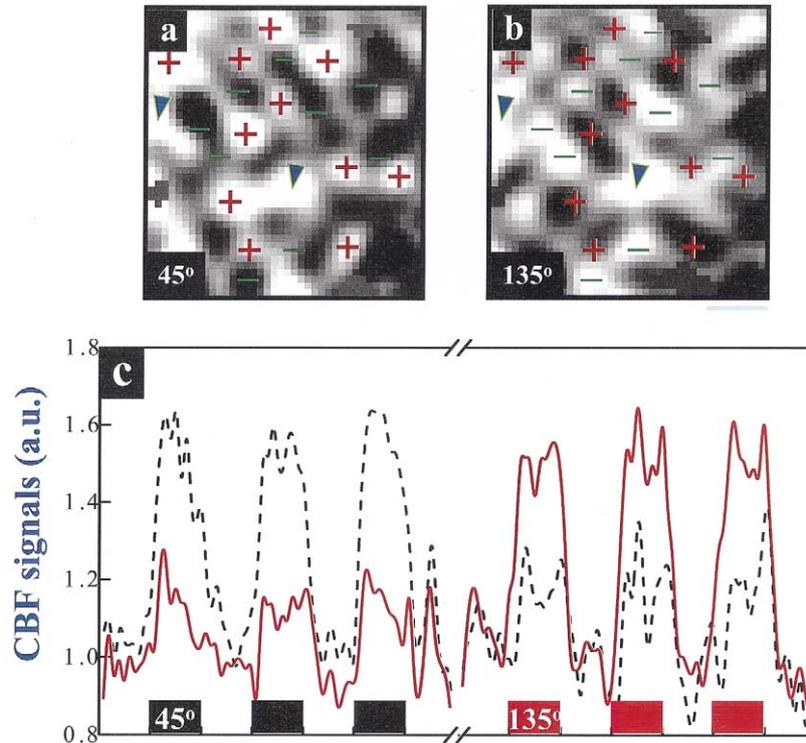


Fig. 5. Activation maps of two orthogonal iso-orientation domains in the cat visual cortex obtained separately with perfusion based mapping. A, T_2^* BOLD vs. Perfusion-based functional maps obtained with a single orientation. Note that while the single condition T_2^* BOLD map is contiguous for one orientation, contrary to what is expected, the corresponding perfusion map yields a patchy image consistent with an iso-orientation domain. B, Perfusion-based images obtained with two orthogonal orientations (45° and 135° gratings), demonstrating their complementarity. Each map was acquired as a single condition map where the grating of one orientation was either moving back and forth (activation condition) or static (control condition). Bottom trace shows the blood flow (perfusion) response to in all the voxels identified as “activated” either by the 45° or 135° gratings for the two orthogonal stimulations. All voxels identified as “activated” for 45° orientation display a large perfusion increase during stimulation by this orientation gratings and show a smaller but detectable perfusion increase in response to the 135° gratings. The opposite of this is observed for the voxels identified as activated for 135° gratings. A marked perfusion increase ($\sim 55\%$) following 45° or 135° stimulus was observed in the regions tuned to these orientations, while the stimulation with the orthogonal orientation lead to a 3.3 ± 0.6 fold smaller perfusion increase in the same region. Adapted from reference [58]

ume of tissue must also increases due to conservation of mass; thus, blood vessel dilation produced by neuronal activation propagates to distant blood vessels [52]. However, long tagging times (~ 2 s) [53] eliminate arterial component from perfusion images. The long tagging times, on the other hand, may lead to the tagged spins appearing in the venous side, thus leading to false activation in the veins. At high magnetic fields, the spin-spin relaxation rate (T_2) of venous blood is very short [30,55] so that this effect can be selectively eliminated by a brief delay after excitations of spins but before image acquisition. Accordingly, perfusion-based fMRI maps have been shown to co-localize with Mn^{+2} uptake [56], a marker of calcium dependent synaptic activity [57].

In the absence of large vessel contributions, perfusion-based fMRI can be used to examine the critical physiological question related to specificity of blood flow increases. Namely, are perfusion changes confined accurately to the region of increased neuronal activity in the spatial scale of columnar organizations? Such a study was performed recently using the iso-orientation columns in the cat visual system [58]. It demonstrated that while perfusion increases

that follow neuronal activation were not “perfectly” localized at the iso-orientation column level, the difference between active and neighboring inactive columns was large and permitted single condition mapping (Fig. 5). This contradicts the concept based on optical imaging data that blood flow responses are not specific in the columnar scale and extend several millimeters beyond the area of activation. However, the MR results are not necessarily contradictory to the optical imaging data themselves. Optical imaging does not measure blood flow (rather, the amount of hemoglobin, which is determined by blood volume and red cell density), and cannot selectively eliminate noncapillary contributions. In addition, the “contrast” in these two measurements (i.e., the *difference* detected between the active and inactive column) are likely to be smaller for CBV, since fractional CBV changes are expected to be smaller than fractional blood flow changes [59]. Recent studies in fact demonstrate that optical imaging based on CBV changes can resolve columnar structures provided large vessel effects are eliminated, in total agreement with the perfusion imaging results [60].

Thus, the explanation for the failure of T_2^* BOLD fMRI

to depict columnar organization in *single-condition mapping*, for having a spatial accuracy in the 3 to 5 mm range (when assessed either by electrophysiology or retinotopy in the human brain) is the presence of large vessel contributions to T_2^* BOLD fMRI. The deoxyhemoglobin changes which are initiated at the point of increased neuronal activity do not remain stationary and propagate into large “draining” vessels. Thus, they appear, incorrectly, as “activation” distant from the actual site of neuronal activity. The extent of this contribution is expected to depend on the magnetic field magnitude because it arises both from blood and from extravascular BOLD effect (e.g., see [61–64]). The extravascular BOLD effect will increase linearly with magnetic field magnitude while intravascular effect diminishes rapidly with magnetic field magnitude beyond about 3 Tesla [55].

The presence of large vessel contribution in T_2^* BOLD fMRI images even at 7 Tesla can be seen if perfusion based images can be compared with T_2^* BOLD fMRI. Both because of the enhanced SNR and the long T_1 , high fields provide advantages for perfusion based functional imaging. In high resolution perfusion images that were obtained at 7 Tesla for the first time [13], the activity was found to be confined to the gray matter ribbon; in contrast, the T_2^* BOLD fMRI images showed highest intensity changes not in the cortex but in the cerebral spinal fluid space (CSF) in the sulcus (Fig. 6). Within the cortical gray matter, there are 50- to 100-micron veins separated by 1 to 1.5 mm that drain the capillaries. These are the blood vessels that can be seen in high resolution T_2^* -weighted images as dark lines traversing the gray matter perpendicular to the cortical surface. These small veins drain immediately into similarly sized or somewhat larger veins on the cortical surface. Since these vessels drain a small territory, the deoxyhemoglobin alterations induced at the active site at the terminal ends of the capillaries are mostly, if not fully, preserved in going from capillaries to these small diameter post-capillary vessels. Consequently, these small intracortical and surface venules and veins must have a significant extravascular effect, one that must exceed the capillary contribution to extravascular BOLD since, unlike the capillaries, they are large enough to be outside the dynamic averaging regime that diminishes the BOLD effect (and, hence, converts it from a T_2^* to an apparent T_2 phenomenon). The small veins located on the cortical surface, are likely the source of intense but false activation seen within sulci in human T_2^* BOLD images at 7 Tesla [13] (and Fig. 6).

These images, faithful to the borders of the perfusion increase without any large vessel contribution, can be obtained also with Hahn Spin Echo (HSE) BOLD at high but *not* low magnetic fields. HSE fMRI responds to apparent T_2 (as opposed to T_2^*) changes originating from diffusion of water in the presence of magnetic field gradients generated in the extravascular space around microvasculature, as well as from exchange of water in and out of red blood cells in blood itself [61,63,65]. The former provides spatial specificity in the 100-micron spatial scale because capillaries are

separated on the average by 25 μm [66]. The blood effect, however, can be associated with both large and small blood vessels and hence degrades spatial specificity of fMRI. However, the apparent T_2 of venous blood decreases quadratically with magnetic field magnitude [67] and is diminished from ~ 180 msec at 1.5 Tesla [68] to ~ 6 msec at 9.4 Tesla [54], significantly smaller than brain tissue T_2 and the TE values that would be used at such field strengths. HSE BOLD signals associated with capillaries are inherently small and undetectable at low magnetic fields. But they are expected to increase supralinearly with magnetic field. They approach potentially useful dimensions at 3 Tesla [69] (provided blood contribution that still exists at this field strength is suppressed) and are large enough to yield microvasculature-based *high* resolution maps even in the human brain at 7 Tesla (Fig. 7) and 9.4 Tesla [54]. As in perfusion imaging, HSE BOLD techniques are expected to yield columnar level mapping. That they are capable of distinguishing among different layers has already been shown, as indicated in Fig. 7. In these HSE fMRI images, the activation follows the cortical gray matter with the largest intensity in the middle layers rather than in the outer surface and in the intrasulcal space. The appearance of the largest intensity in the middle layers (rather than the cortical surface) is expected since the inputs from the lateral geniculate nuclei terminate in layer 4. This is also the layer with the highest capillary density and cytochrome oxidase activity.

In principle, it is possible to obtain functional images with better spatial specificity than perfusion if one can map the changes in cerebral oxygen consumption rate (CMRO_2). CMRO_2 increases must occur through increased activity of the mitochondria in the “activated” neurons. The mitochondrial density is high in the synapse rich regions, predominantly in the dendrites [70]. It might be possible to image CMRO_2 increases directly using the magnetic isotope of oxygen (^{17}O) (e.g., [71–73] and references therein). CMRO_2 increases can also be mapped indirectly through its perturbations of deoxyhemoglobin content. It has been shown that optical imaging of intrinsic signals in the cat and monkey visual cortex display a transient and small signal change ascribed to the deoxyhemoglobin increase before the onset of a larger increase in blood flow [74–78]. In this explanation, it is the lag in the blood flow response which provides the *temporal opportunity* to detect the deoxyhemoglobin increase. The optical imaging data also showed that this deoxyhemoglobin increase yielded better maps of iso-orientation or ocular dominance columns than the signals associated with the hyperoxygenated state. Deoxyhemoglobin increase, however, can only be co-localized with elevated CMRO_2 at early time points after the onset of CMRO_2 elevation. Even if subsequent blood flow enhancement did not occur, and the deoxyhemoglobin increase was the only response detected, this deoxyhemoglobin increase would co-localize with enhanced neuronal activity initially, but later would propagate down the vasculature and appear in the venous system as discussed previously.

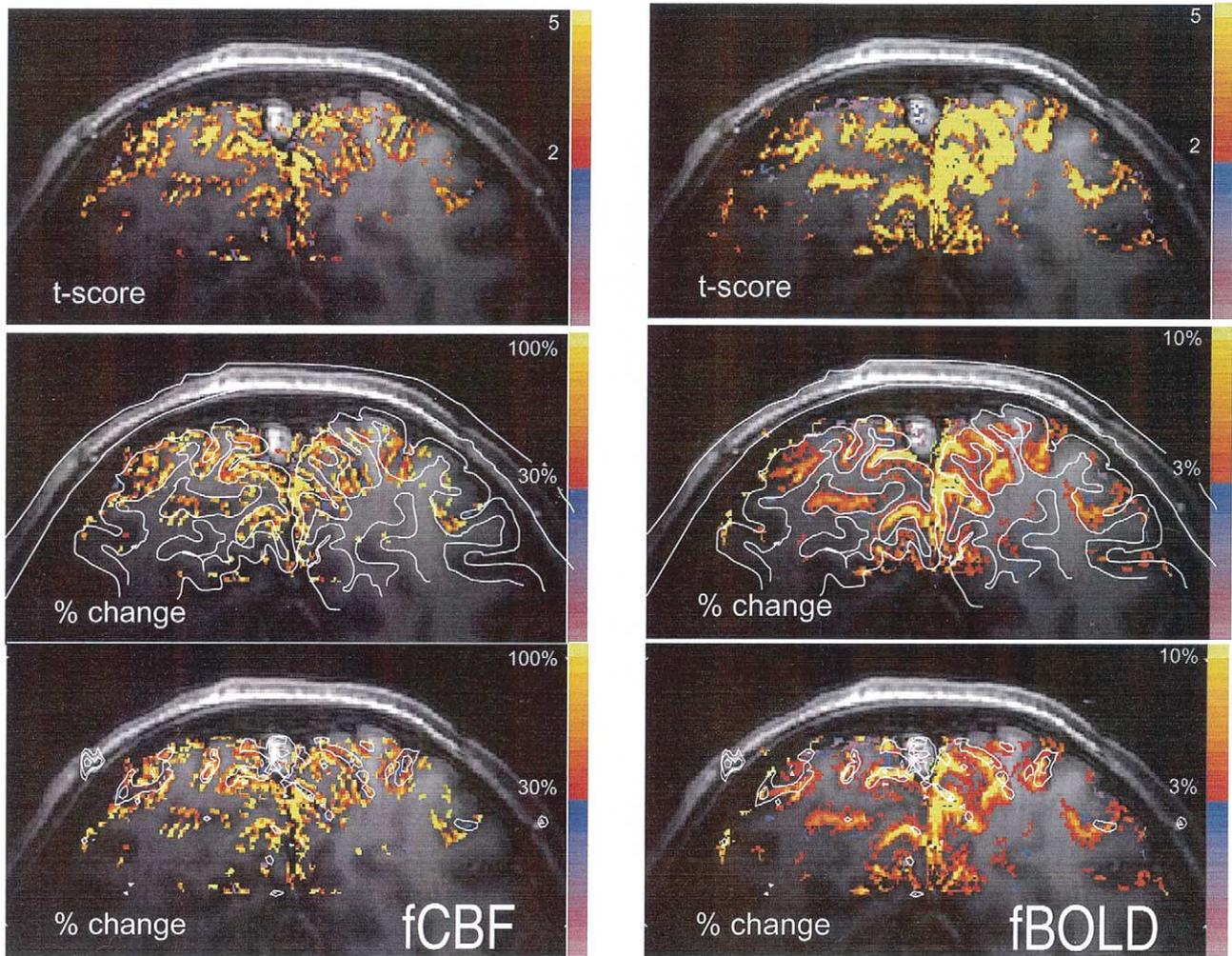


Fig. 6. 7 Tesla functional perfusion (left column) and T_2^* BOLD (right column) maps obtained in the human brain with $0.9 \times 0.9 \times 3 \text{ mm}^3$ spatial resolution during visual stimulation. The T_2^* BOLD acquisition was interleaved with the acquisition of the fCBF maps. Results represent a t -test analysis with $p < 0.05$. Percent changes range from 30% to over 100% for fCBF and from 3% to over 10% for fBOLD, respectively. Anatomic contours of the gray matter are overlaid onto the %change map in the middle row.

The transient deoxyhemoglobin increase would yield a negative signal change in a BOLD-weighted MR image. Such a “dip” has been observed in human fMRI experiments [31,79–84] and in animal models [36,37,85]. However, both in optical imaging and fMRI literature, the observation of this “dip” has been controversial. The source of this controversy in the early human fMRI studies has been the failure of several laboratories to detect this effect at 1.5 Tesla studies. However, this is a small effect at low magnetic fields but was shown to increase substantially with magnetic field magnitude [31] to yield easily detectable signals at 7 Tesla. Recently, in intra-operative studies in the human brain, the functional maps obtained from this early “dip” was also detected and shown to co-localize with electrical recordings, whereas T_2^* BOLD maps did not [84]. In anesthetized animal models, this small dip has not been highly reproducible presumably because of the perturbations of anesthesia on the animal physiology. When observed, it was found to yield accurate maps at the iso-

orientation columnar level [36,37]. However, this early negative response corresponding to deoxyhemoglobin increase is a very small effect. Furthermore, in order to obtain accurate maps from it, one must restrict the observation window to early time points where the signals are even smaller. Thus, as a general high resolution MR imaging approach, it suffers from low contrast-to-noise.

3. Imaging using low gyromagnetic nuclei

Low gyromagnetic nuclei can in principle be used to obtain unique biologic information. Atoms with magnetic nuclei such as phosphorus, oxygen, carbon (with magnetic isotopes ^{31}P , ^{17}O , ^{13}C) all appear in abundance within cells. Ions such as sodium and potassium (with magnetic isotopes ^{23}Na , ^{39}K) are critical in cellular function. However, these low gyromagnetic nuclei suffer from limitation in sensitivity. Not only is the inherent MR sensitivity for detection of

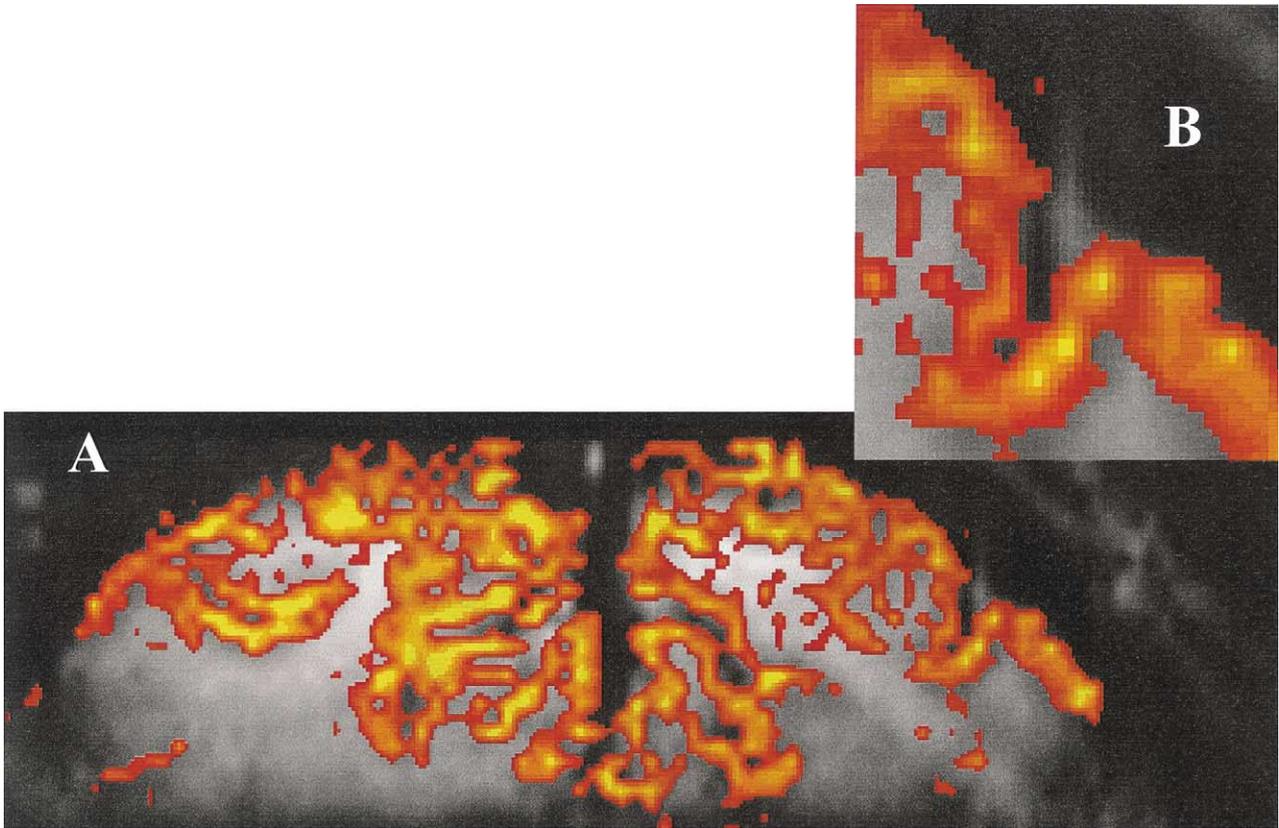


Fig. 7. Spin-echo images ($0.5 \times 0.5 \times 2 \text{ mm}^3$) of the visual activation in the human visual cortex obtained at 7 Tesla. The data demonstrate that high spatial resolution images can be obtained robustly in the human brain at 7 Tesla using the normally weak but more accurate Hahn-Spin Echo Imaging signals. The insert (B) shows an expanded view around a sulcus demonstrating that the highest intensity HSE BOLD activation is in the middle layers (unlike T_2^* BOLD) and the CSF space in the sulcus is free of high intensity changes coupled to “false” activation, unlike T_2^* BOLD (as shown in Fig. 6, right column). Note also the absence of BOLD signal intensity changes between the two hemispheres, again, unlike the T_2^* BOLD images shown in Fig. 6, right column. Single Shot Spin Echo Planar Imaging with field of view reduction to $3.2 \text{ cm} \times 12.8 \text{ cm}$ (64×256) (partial Fourier) TE/TR: 70 msec/2s. From [123].

these nuclei low because of their small magnetic moments, their abundance in the cells as a single chemical species (i.e., with one chemical shift) is also low. Therefore, sensitivity gains that can be realized with increasing magnetic fields become critically important for these nuclei.

At high frequencies attained by protons at magnetic fields like 7 or 9.4 Tesla, signal-to-noise ratio dependence on magnetic field becomes complex and approximately linearly dependent on magnetic field magnitude. However, for low frequencies or for nonconductive samples, the magnetic field dependence of SNR can be approximately quadratic and is expressed by the relationship

$$\text{SNR} \propto B_0^\beta (T_2^*/T_1)^{1/2}, \tag{1}$$

where, based on theoretical considerations, the constant β was suggested to be $7/4$ [86,87]. Motivated by the SNR gains expected from the availability of high field magnets, in recent years, we have examined the possibility of using ^{17}O for monitoring oxidative metabolism in intact animals and ultimately humans at 7 and 9.4 Tesla at resonance frequencies of 40 and 54 MHz, respectively.

3.1. Oxygen 17 imaging

^{17}O spin is the only stable oxygen isotope having a magnetic moment that can be detected by the NMR approach. It has a spin quantum number ($I = 5/2$) greater than one half and possesses an electric quadrupole moment. This is the dominant source of T_1 and T_2 relaxation [88] for this nucleus. In the case of water molecule, for which the extreme narrowing limit (i.e., $\tau_c \omega \ll 1$, where τ_c is the rotational correlation time and ω is Larmor frequency in radians/sec) is applicable (except for the bound water) the values of T_1 and T_2 can be estimated by using the equation

$$\frac{1}{T_2} \cong \frac{1}{T_1} \cong \frac{3}{40} \left(\frac{2I + 3}{I^2(2I - 1)} \right) \left(1 + \frac{\eta^2}{3} \right) \left(\frac{\pi e^2 Q q}{h} \right)^2 \tau_c \tag{2}$$

where the term of $(\pi e^2 Q q / h)$ is the quadrupole coupling constant and is equal to 6.7 MHz (or $4.21 \cdot 10^7$ radian/s) for the ^{17}O nucleus, η is an asymmetry parameter ($0 \leq \eta \leq 1$) [88].

Because both η and τ_c are independent of the magnetic field strength, T_1 and T_2 (or T_2^*) for the ^{17}O spins should be

field independent. If the τ_c value of 8.5×10^{-12} s is used for bulk water at 298 K [89], then estimated T_2 and T_1 should be between 2.8 ms (for $\eta = 0$) and 2.1 ms (for $\eta = 1$). The actual ^{17}O T_2 value should be smaller than the ^{17}O T_1 value due to the significant contributions of the ^{17}O - ^1H scalar coupling and the proton chemical exchange to the ^{17}O T_2 relaxation processes [90]. Consistent with these expectations, in experiments conducted at 4.7 and 9.4 Tesla, the ^{17}O relaxation times in water were found to be short, similar, and field independent for the rat brain ($T_2 = 3.03 \pm 0.08$ ms, $T_1 = 4.47 \pm 0.14$ ms at 4.7 Tesla; $T_2 = 3.03 \pm 0.09$ ms, $T_1 = 4.84 \pm 0.18$ ms at 9.4 Tesla), as well as for the saline solution at the room temperature ($T_2 = 4.28 \pm 0.08$ ms, $T_1 = 6.59 \pm 0.10$ ms at 4.7 Tesla; and $T_2 = 4.09 \pm 0.06$ ms, $T_1 = 6.52 \pm 0.12$ ms at 9.4 Tesla) [91].

The ^{17}O NMR sensitivity was evaluated at 9.4 Tesla and 4.7 Tesla in the rat brain and in saline solution [91]. A consistent SNR gain of approximately fourfold was observed at 9.4 Tesla as compared to 4.7 Tesla, indicating an approximately theoretically maximum power dependence of SNR on B_0 . These experimental results demonstrate the significant advantage provided by high field strength for direct detection of ^{17}O NMR signal. This trend for increasing ^{17}O NMR sensitivity is expected to hold beyond 9.4 Tesla.

With the dramatic sensitivity gains realized at 9.4 Tesla, it is possible to obtain 3-D ^{17}O imaging and utilize it for calculation of oxygen consumption rates. The dynamic change in cerebral H_2^{17}O concentration during an inhalation of $^{17}\text{O}_2$ is determined by three parallel processes: 1) oxygen consumption generating the metabolic H_2^{17}O from ^{17}O in the brain; 2) H_2^{17}O washout from the brain; and 3) flow recirculation bringing extra H_2^{17}O into the brain. The mass balance equation of labeled H_2^{17}O concentrations during an inhalation of $^{17}\text{O}_2$ is given by

$$\frac{dC_b(t)}{dt} = 2\alpha f_1 \text{CMRO}_2 + m \text{CBF} \left(f_2 C_a(t) - \frac{n C_b(t)}{\lambda} \right) \quad (3)$$

where $C_a(t)$, $C_b(t)$ are the time-dependent H_2^{17}O concentrations in excess of the natural abundance H_2^{17}O concentration level in the arterial blood, and the brain tissue, respectively; α is the ^{17}O enrichment fraction of inhaled $^{17}\text{O}_2$ gas. The parameters C_b , C_a can be measured by ^{17}O NMR and they can be calibrated using the natural abundance H_2^{17}O concentration (20.35 μmole per gram brain water for brain tissue, and $\mu\text{mol per gram blood water}$ for blood, calculated from natural abundance H_2^{17}O enrichment of 0.037% and the molecular weight of $\text{H}_2^{17}\text{O} = 19.0$). The mass balance equation for C_b is written in terms of preferred units $\mu\text{mol}/(\text{g brain water})$ for $C_b(t)$, $\mu\text{mol}/(\text{g blood water})$ for $C_a(t)$. The unit used most commonly for CMRO_2 is $\mu\text{mole}/\text{min}/(\text{g brain tissue})$ and is retained. This dictates the use of

two unit conversion factors, $f_1 = 1.266$ and $f_2 = 1.05$, to achieve consistency of units among all parameters used [92,93]. The equation above assumes water in the brain tissue is in equilibrium with water in the venous blood (i.e., fast water exchange across capillaries); thus, $f_2 C_v(t) = C_b(t)/\lambda$ where C_v is the H_2^{17}O concentrations in excess of the natural abundance H_2^{17}O concentration level in venous blood and λ is the brain/blood partition coefficient (≈ 0.90) in units of $(\text{ml blood})/(\text{g brain tissue})$ [94]. The parameter m is a correction factor accounting for the limited permeability of water because water is not totally freely diffusible across the brain-blood barrier; in the CBF range attained in these studies, m is 0.84 [95]. The constant n is another correction factor that accounts for the restriction on the permeability of the H_2^{17}O generated through oxidative metabolism in the mitochondria (“metabolic” H_2^{17}O) during an inhalation of $^{17}\text{O}_2$. This additional restriction is included in our modeling because we have observed that the washout rate of the metabolic H_2^{17}O after the cessation of $^{17}\text{O}_2$ inhalation was significantly slower than the washout rate of the H_2^{17}O that permeates brain tissue subsequent to a bolus injection of H_2^{17}O through the internal carotid artery in the rat [96].

The CMRO_2 value can, therefore, be precisely calculated from ^{17}O imaging studies using the complete model if all parameters of $C_b(t)$, CBF, n and $C_a(t)$ can be determined experimentally for the same brain. However, experiments have shown that at least in the rat model, $C_a(t)$ can be approximated as a linear function of the inhalation time (i.e., $C_a(t) \approx At$, where A is a constant and t represents time) or even ignored for short inhalation times [91].

A fundamental limitation of spatial resolution using the ^{17}O imaging approach will be water diffusion in the brain since water is usually assumed freely diffusible. The apparent diffusion constant of water in the rat brain has been measured to be $\sim 10^{-3}$ mm^2/s [97] with small direction-dependent variations due to anisotropy. The average “lifetime” of a H_2^{17}O molecule in the brain tissue after it is formed and before it is washed out by blood flow is given by λ/CBF which is ~ 1 min since $\text{CBF} \sim 1$ $\text{ml}/\text{g}/\text{min}$. Thus, the diffusion distance during this lifetime is ~ 0.25 mm. Therefore, major limitation on spatial resolution for imaging CMRO_2 will be determined by the ^{17}O NMR sensitivity and not the diffusion of water.

We have recently investigated the feasibility for developing the ^{17}O NMR approach for imaging CMRO_2 in a small animal model during a brief inhalation of $^{17}\text{O}_2$ gas at 9.4 Tesla [93]. Fig. 8A demonstrates stacked plots of ^{17}O spectra of cerebral H_2^{17}O from one representative voxel (0.1 ml volume) before (natural abundance), during and after a 2-min inhalation of $^{17}\text{O}_2$ (72% ^{17}O enrichment) gas [93]. The ^{17}O signal intensity of cerebral H_2^{17}O is characterized by three distinct phases: 1) constant before the $^{17}\text{O}_2$ inhalation; 2) approximately linear increase during the $^{17}\text{O}_2$ inhalation; and 3) approximately exponential decrease after the cessation of $^{17}\text{O}_2$ inhalation and approaching a new steady concentration within a short recovery time (<10

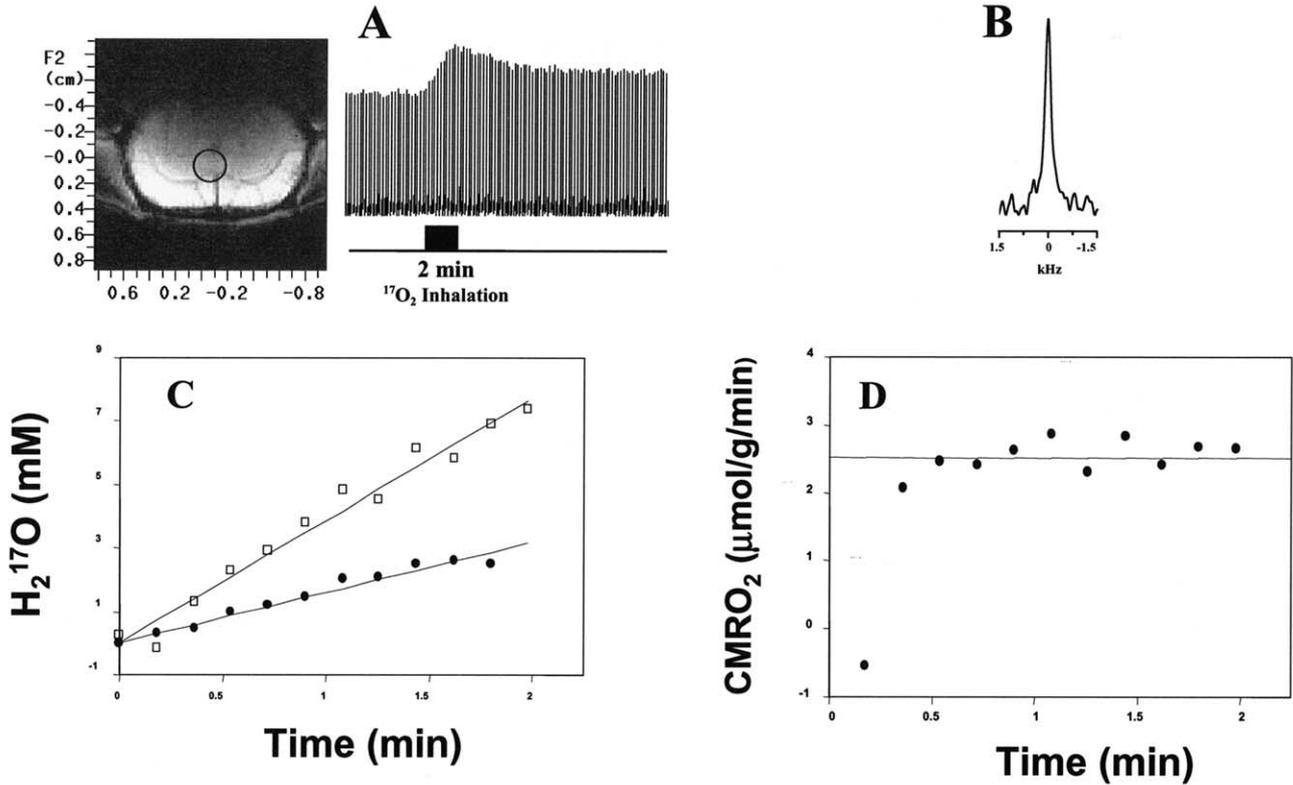


Fig. 8. A, Stacked plots of the cerebral $H_2^{17}O$ spectra from one representative voxel (0.1 ml voxel size) as indicated by the circle in the anatomic image (left insert) acquired before (natural abundance), during (as indicated by the dark bar under the stacked plots) and after a 2-min $^{17}O_2$ inhalation. ^{17}O spectrum of natural abundance $H_2^{17}O$ in the rat carotid artery blood ($\sim 7 \mu L$) obtained using the implanted RF coil before inhalation of $^{17}O_2$ (B) and time course (C) of ^{17}O MR signals of $C_a(t)$ in one carotid artery (circle symbol) and $C_b(t)$ from a representative voxel in the rat brain (square symbol) during inhalation of $^{17}O_2$. D, Plot of the calculated $CMRO_2$ values using the complete modeling as a function of inhalation time.

min). The ^{17}O NMR sensitivity achieved at 9.4 Tesla makes it possible to distinguish the different phases and to quantify $C_b(t)$.

The most challenging measurement faced by the complete model for calculating $CMRO_2$ is to the experimental determination of $C_a(t)$, the concentration of $H_2^{17}O$ in the feeding artery, especially when a small animal model is used. In order to accomplish this, we have designed an implanted ^{17}O RF coil that permits the continuous measurement of $C_a(t)$ in a rat carotid artery [93]. This coil is based on a modified solenoid coil design combined with an RF shield. The RF shielding ensures that the NMR signal detected by the implanted coil originates only from the arterial blood ($\sim 7 \mu L$) without contamination from surrounding tissues. Therefore, additional spatial localization was not necessary for determining $C_a(t)$. This setup allowed simultaneous measurements of both $C_a(t)$ and $C_b(t)$ using two independent coils and receiver channels with the same temporal resolution (11s). Fig. 8B illustrates the natural abundance $H_2^{17}O$ spectrum detected from the blood in the rat carotid artery. Fig. 8C plots the results of simultaneous measurements of $C_a(t)$ change in one carotid artery and $C_b(t)$ change in a voxel located in the brain from the same rat during an inhalation of $^{17}O_2$. The results indeed demonstrate a linear relation between $C_a(t)$ and the inhalation time

t. Fig. 8D demonstrates one example of $CMRO_2$ calculation as a function of inhalation time from a *single* voxel in the rat brain from such an ^{17}O experiment. It is evident that the $CMRO_2$ value is independent of the inhalation time if the initial two points are excluded, indicating that the tissue concentration of ^{17}O labeled oxygen molecule rapidly approaches the K_m of cytochrome Oxidase for oxygen after inhalation of ^{17}O gas commences.

The averaged $CMRO_2$ and CBF values in the rat brain from seven measurements (five rats) were $2.19 \pm 0.14 \mu mol/g/min$ and $0.53 \pm 0.07 ml/g/min$, respectively, under conditions of α -chloralose anesthesia [93]. This is in excellent agreement with what can be calculated from an autoradiographic measurement [98] of cerebral metabolic rate of glucose (CMR_{glc}) ($\sim 0.37 \mu mole/g/min$) using the $CMRO_2/CMR_{glc}$ ratio of 5.5–6.0 [99]. In this autoradiographic study, both the anesthesia conditions and the CBF attained was similar to that in our own.

The $CMRO_2$ measurements described in this chapter involve two invasive procedures, one for determining $C_a(t)$ by means of the implanted ^{17}O RF coil, the second for measuring CBF by an intra-arterial catheter and bolus injection of $H_2^{17}O$. The CBF measurement can be performed noninvasively using the arterial spin tagging MR approaches. The use of the implanted RF coil, however, is not

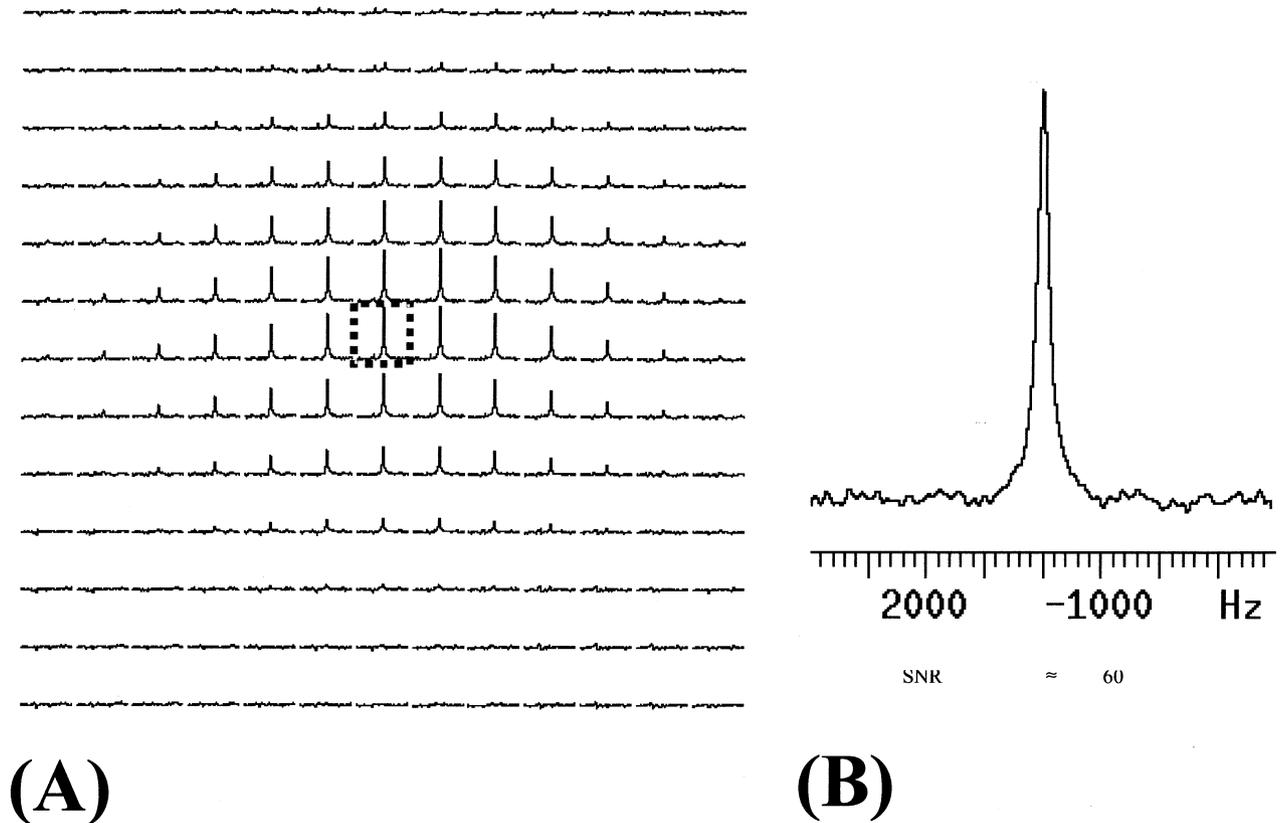


Fig. 9. A, ^{17}O CSI of natural abundance H_2^{17}O in the coronal orientation, ^{17}O acquired from the human visual cortex at 7 Tesla. B, One representative ^{17}O spectrum from one central voxel as shown in A with 8.5 s of image acquisition time and 6.6 ml voxel size.

suitable for routine measurements of CMRO_2 using small experimental animal models (e.g. rat or mouse), or for human applications. Therefore, it is important to further explore the feasibility of the ^{17}O NMR approach for reliably imaging CMRO_2 without resorting to invasive $C_a(t)$ and/or CBF measurements. Recently, we have investigated and showed this feasibility for imaging CMRO_2 in the rat brain using simplified models, which only require noninvasive measurements of the cerebral H_2^{17}O concentration (i.e. $C_b(t)$), yielding similar CMRO_2 results with that obtained using the complete model accounting for all parameters [100]. This work opens the possibility to establish a completely noninvasive ^{17}O NMR approach for CMRO_2 imaging. However, this possibility needs further validations in different physiological conditions and species. This is, in particular, interesting and important for human applications, where elimination of invasive procedures is desirable if not absolutely necessary.

The success of high-field ^{17}O NMR for imaging CMRO_2 in the human brain again relies on the ^{17}O NMR sensitivity. The sensitivity loss caused by the use of large RF coil size could be partially compensated by increasing image voxel size in the humans, which have a much large brain in comparison with most animals. Recently, we have studied the in vivo NMR sensitivity for acquiring 3-D ^{17}O images of natural abundance H_2^{17}O in the human visual cortex at 7

Tesla with a temporal resolution of 8.5 s and a voxel size of 6.6 ml. Fig. 9A demonstrates the results obtained from a representative subject showing one coronal ^{17}O CSI of natural abundance H_2^{17}O (extracted from 3-D ^{17}O CSI dataset). The CSI plane was located in the human visual cortex. Fig. 9B displays a representative ^{17}O spectrum from one central voxel showing the resonance peak of natural abundance H_2^{17}O in the human visual cortex and excellent SNR ($\approx 60:1$). This is better than that obtained from the rat brains at 9.4 Tesla ($\sim 40:1$) albeit with a larger absolute voxel size but not relative to the size of brain [91,93]. Furthermore, the dynamic concentration change of metabolic H_2^{17}O during an $^{17}\text{O}_2$ inhalation is expected to be similar between the awake human brain and the rat brain anesthetized with α -chloralose because of similar values for both CMRO_2 ($2.2 \mu\text{mol/g/min}$ in the anesthetized rat brain versus $1.7 \mu\text{mol/g/min}$ in the awake human visual cortex) and CBF (0.53 ml/g/min in the anesthetized rat brain versus 0.54 ml/g/min in the awake human visual cortex) [93,101]. This comparison indicates that conditions in the rat brain in our studies and the awake human brain are comparable relative to the parameters that impact determination of CMRO_2 from the ^{17}O data. Thus, it will be possible to achieve 3-D imaging of CMRO_2 in the human brain at 7 Tesla or greater magnetic fields. Realization and utilization

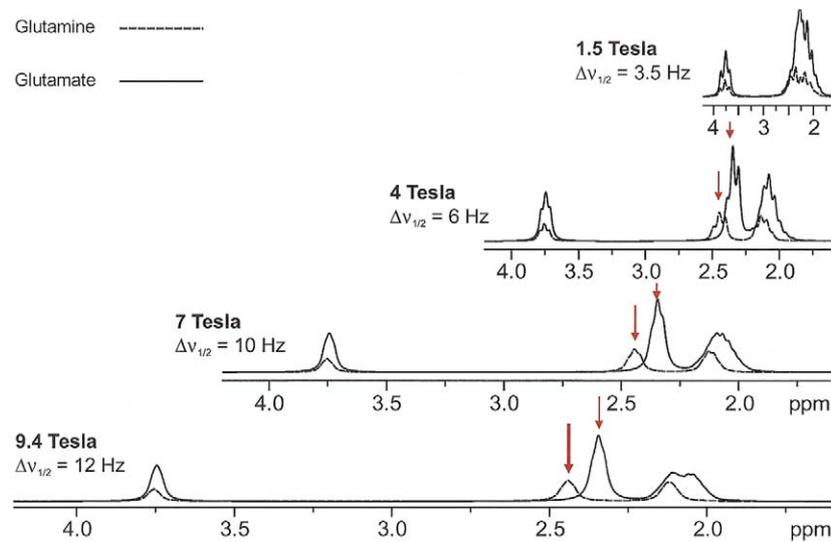


Fig. 10. Simulated ^1H spectrum of glutamate and glutamine at four different field strengths assuming that the line widths increase linearly with the magnetic field magnitude. Adapted from [103]

of this potential could have an important impact on studying bioenergetics in the human brain.

4. Spectroscopy at high magnetic fields

SNR gains somewhere between linear as observed for the ^1H spins and near quadratic as seen for ^{17}O nucleus will also be valid for spectroscopy studies that often utilize ^1H , ^{31}P , ^{13}C nuclei. Spectroscopy studies with these nuclei, however, will also benefit significantly from unique advantages besides improved SNR at higher field strengths. Higher B_0 fields yield improved resolution due to chemical shift and spectral simplifications of coupled spin systems.

In ^1H spectra obtained from the human brain, glutamate is resolved from glutamine partially at 4 T but not at 1.5 T; these resonances are fully resolved at 9.4 Tesla [102]. For coupled spins, this resolution is attained even when the linewidths increase linearly with the magnetic field. This is because often the individual resonances of a multiplet are not resolved fully from each other in *in vivo* spectra and what is critical is the resolution of the entire multiplet from another multiplet. Even when the linewidths increase linearly with magnetic field, the effective width of the multiplet does *not* increase linearly because the splitting in the multiplet (i.e., J-coupling) is field independent. The resolution gained in this fashion is illustrated in Fig. 10 using simulation of glutamate and glutamine. The result of sensitivity and resolution gain is shown for human brain at 7 Tesla in Fig. 11.

An excellent example of the resolution afforded by high fields in ^1H spectroscopy is the detection of the 5.23 ppm resonance of α -glucose at 4 T (and higher fields) despite its proximity to H_2O [102,103], providing for the first time the ability to measure cerebral glucose quantitatively by ^1H

MRS without complications from overlapping resonances. All previous ^1H MRS studies concerned with MR detection of glucose at lower fields (e.g., [104–106]) have relied on the 3.4 ppm peak of glucose which suffers from significant overlap from other resonances such as taurine in this very crowded region of the brain spectrum. As such, the 3.4 ppm resonance cannot and should not be quantitatively used for measuring absolute glucose content in the human brain. The 5.23 ppm peak of glucose, on the other hand, is isolated and does not suffer from overlap. However, this resonance is too close to the very large H_2O peak to be detectable at the lower field strength because selective suppression of the latter without affecting the former has not been possible at the lower magnetic fields. The increased chemical shift resolution at 4 T or higher fields, however, permits selective suppression of the H_2O resonance and detection of the 5.23 ppm peak of α -glucose C1-H.

^{31}P NMR studies have been used both in animal models and in human studies to examine tissue bioenergetics, including the bioenergetics of increased neuronal activity. These studies have so far relied on detection of adenosine triphosphate (ATP), inorganic phosphate (Pi), phosphocreatine (PCr). Significant gains are realized at the higher fields for this nucleus as shown in Fig. 12. These data obtained with rats, document better than linear gains in SNR in spatially localized ^{31}P spectra (Fig. 12); such gains permit investigations of bioenergetics in primate brain with greater accuracy and spatial resolution, and have been utilized in the human brain for evaluating pathology and brain activation [107]. Fig. 13 demonstrates ^{31}P spectra from the human brain at 7 Tesla. The SNR and chemical shift resolution available in this 7 T spectrum has not been attainable at lower fields and suggests that numerous experiments that have provided unique biochemical information at very high fields

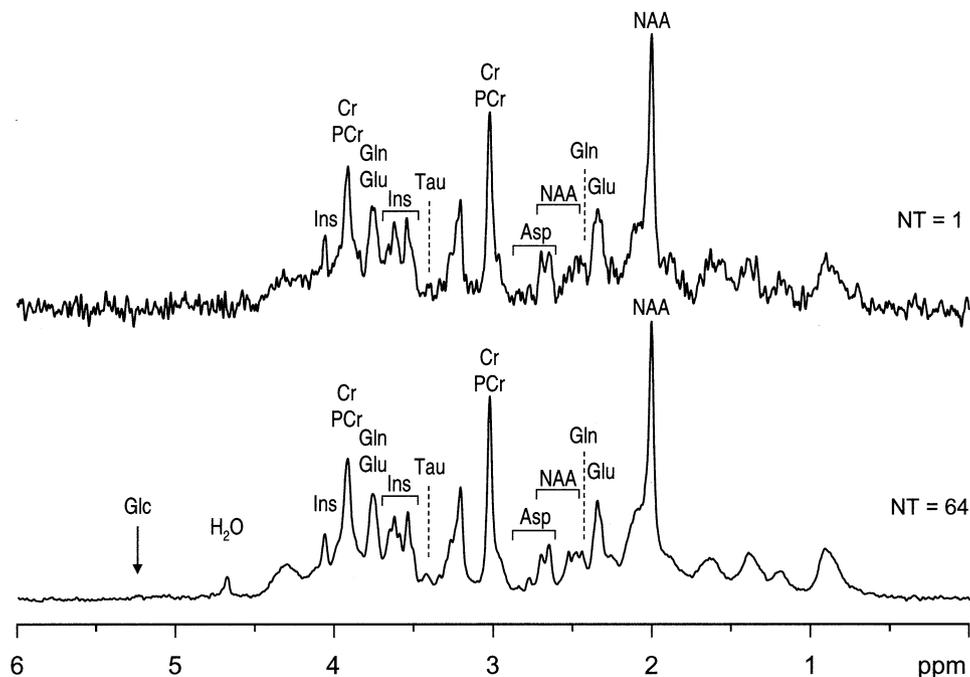


Fig. 11. ^1H NMR spectrum of the human brain at 7 T acquired over the occipital lobe, largely over gray matter, using STEAM ($TE = 6$ ms, $TR = 5$ s, $VOI = 8$ ml) for single shot ($N = 1$) and for 64 averages.

but in animal models will also be feasible for human brain for the first time.

Spatially localized ^{13}C MRS has been employed to measure aspects of brain function and neurotransmission directly. Recent efforts have focused on understanding the coupling between cellular bioenergetics and neuronal activity (e.g. [108–115]). If ^{13}C -enriched glucose is given to a living organism, the ^{13}C label is incorporated through metabolism into several positions in many different metabo-

lites. Of interest is the incorporation of label into the intermediates of tricarboxylic acid (TCA) cycle, which generates reducing equivalents utilized in oxygen consumption. The label is then transferred into the amino acids glutamate, aspartate and glutamine. The ability to monitor the labeling of intracellular compounds in intact cells was first demonstrated using *E. coli* [116]. Today, such highly specific data can be obtained in human and animal brains from relatively small volume elements. Fig. 14 illustrates 9.4 Tesla spectra

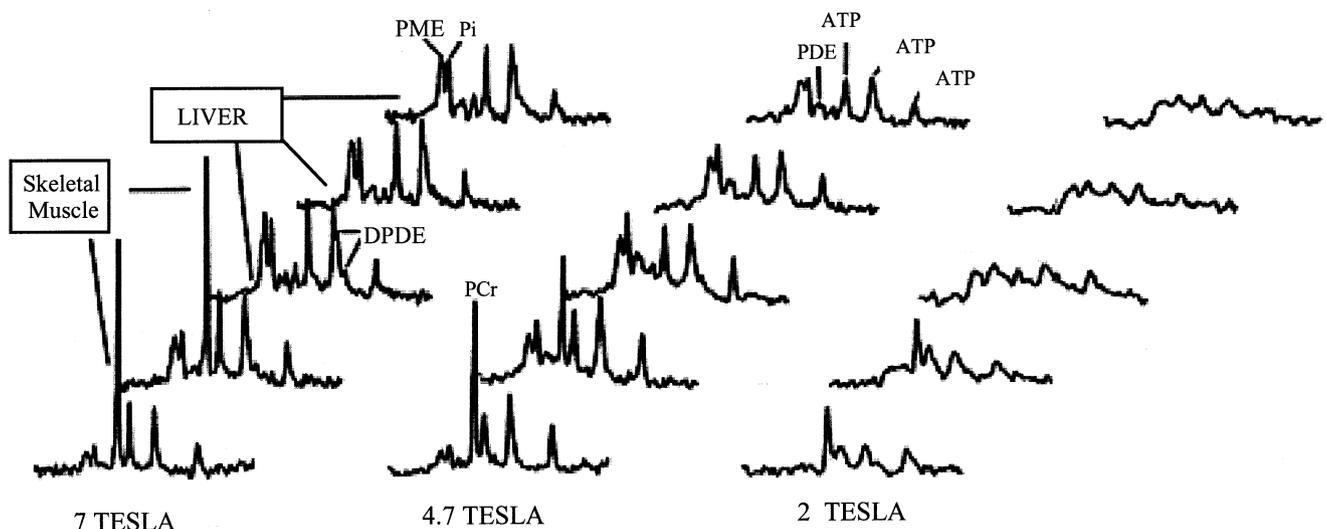


Fig. 12. 2, 4.7, 7 Tesla ^{31}P spectra acquired in five voxels over the abdomen of the same intact, reversibly anesthetized rat. These spectra were obtained using identical techniques, voxel volumes, number of scans, interpulse delay, identical total data acquisition time. Data provided by Dr. S. Weisdorf, M. Garwood, K. Hendrich, and K. Uğurbil.

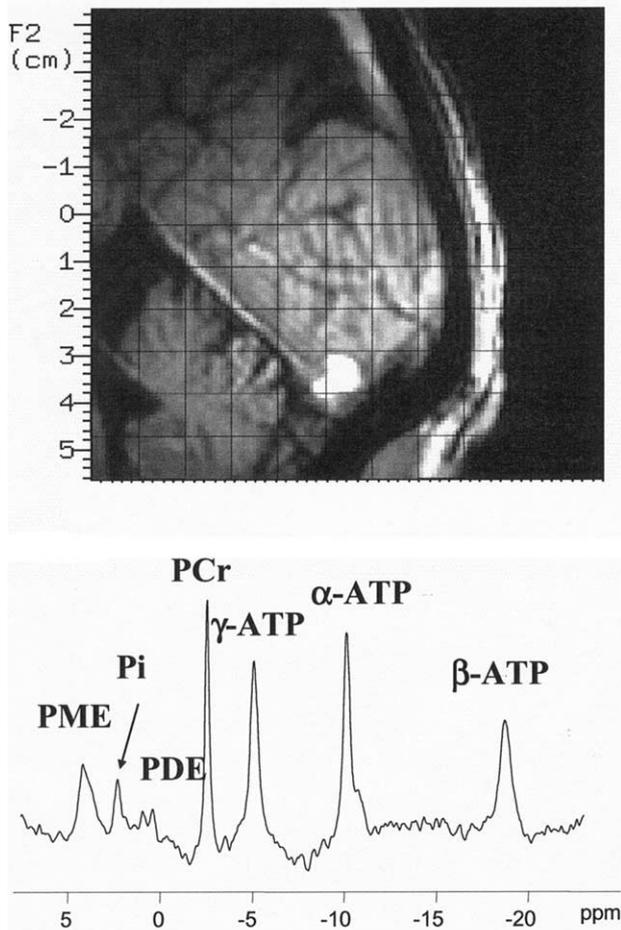


Fig. 13. 7 Tesla human brain ^{31}P MR spectra. Localized ^{31}P spectra were acquired with CSI (TR = 1.5 s and a 200 μs hard pulse for spin excitation). Spectra shown were acquired with 3-D CSI with a voxel size of 1.95 ml and an acquisition time of 7.85 min. Adapted from [124].

from the rat brain, before and after infusion of ^{13}C labeled glucose.

In the the human brain, ^{13}C spectra obtained at 4 Tesla in our laboratory illustrated the detection of many labeled compounds from a relatively small, localized region in the visual cortex, and allowed the first time measurement of neurotransmission rate in the human visual cortex [113]. A subsequent effort on humans at 2.1 Tesla [114] was able to report basically the same results, but used at least three-fold larger volumes (144 ml vs. 45 ml) and longer data acquisition times for signal detection due to the lower field employed. In these studies, it is possible to extract information from the compartmentalized metabolic pathways. Most important is the glutamate-glutamine cycling between glia and the neurons, the two major cell types that are present in the brain: Glutamate is the major excitatory neurotransmitter. Once it is released into the synaptic cleft and binds the post-synaptic receptors during neurotransmission, it is scavenged rapidly by nearby glial processes [117,118]. Glutamate is then converted to glutamine by glutamine-synthase, which is present only in the glia [119]; glutamine is subse-

quently transported to the neurons, converted to glutamate to replenish vesicular glutamate. Thus, glutamine labeling in ^{13}C experiments can in principle occur predominantly through glutamate release through neurotransmission and hence reflect the kinetics of neurotransmission. Therefore, from such ^{13}C studies, it is possible to calculate rates of glutamatergic neurotransmission, TCA cycle turnover, cerebral oxidative glucose utilization (CMRglu) and even break down the latter into glial and neuronal contributions. Most important finding from our studies is the conclusion that neurotransmission rate equals $\sim 60\%$ of total oxidative glucose consumption rate in the awake human brain and that there is significant oxidative ATP production rate in the glia, apparently in contradiction with earlier conclusions reached from experiments on the anesthetized rat brain data [120].

Such detailed studies have not yet been applied to investigate neuronal *activation* in the human or animal brains due to limitations in signal-to-noise ratio of the measurements. However, a subset of the data in the direct ^{13}C measurements can be acquired with indirect detection with higher sensitivity using the protons attached to the ^{13}C nuclei, permitting calculations of changes in TCA cycle turnover and oxidative CMRglu during neuronal activation. Such an experiment was performed at 4 Tesla using hemifield visual stimulation. Spectroscopic data during ^{13}C infusion was used to obtain glutamate labeling kinetics together with CBF and BOLD fMRI from an activated region in one hemisphere while the other hemisphere served as control [108]. Hemifield visual stimulation selectively activates the primary visual cortex of the contralateral hemisphere, thus approximately half the primary visual cortex volume that normally would be engaged during full field stimulation. Spectra were acquired from two volumes positioned so that only one of the two covers the “activated” region within one hemisphere while the other covers the analogous but non-activated region within the other hemisphere. These data put an upper limit on the increase in CMRO_2 of 30% as opposed to larger increases in CBF, supporting the concept that CMRO_2 is not stoichiometrically coupled to increases in CBF and CMRglu. This measurement is significant because of the controversy surrounding the coupling of oxidative metabolism and neuronal activity. It is well known that under resting conditions, the cerebral metabolic rate of glucose consumption is well coupled to CMRO_2 as well as to cerebral blood flow in the human brain ([121] and references therein), the glucose metabolism is almost completely through oxidation. However, this appears not to be the case during increased neuronal activity. Based on PET studies, the increases of CMRO_2 (0–5%) were found to be much less than the elevation in CBF and CMRglc (40–51%) during visual and somatosensory stimulations [27,122]. This early PET result remains to this day highly debated, with the intensity of debate having increased recently in view of its significance in understanding the BOLD response that has come to play such a prominent role in neuroscience research. The difficulties and complexities as-

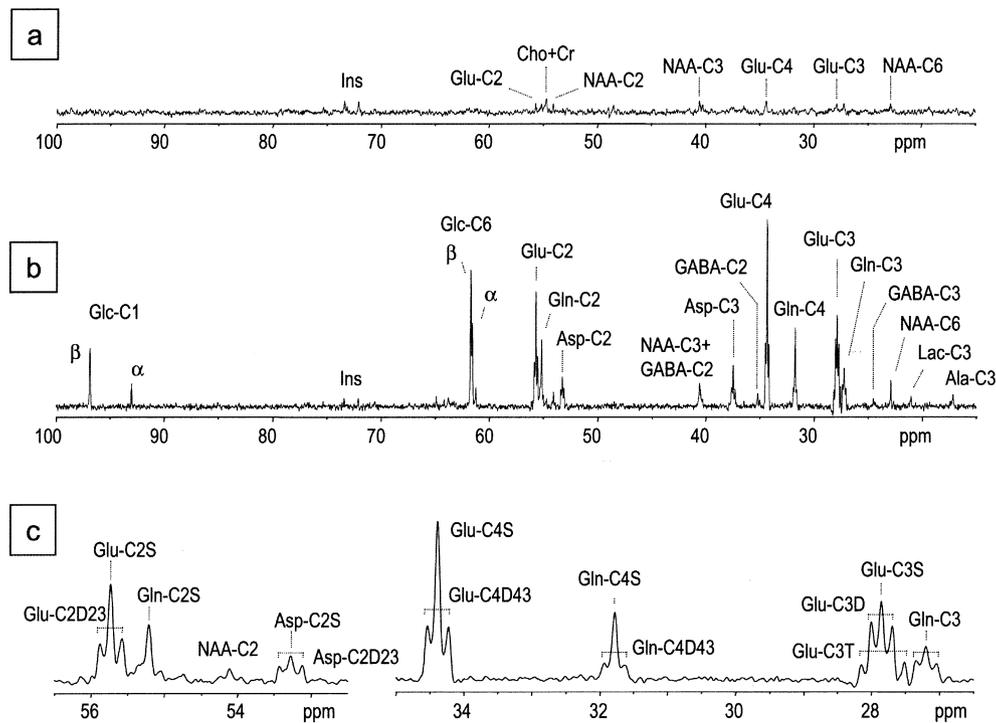


Fig. 14. In vivo ^{13}C spectra from a 400- μl volume in the rat brain, acquired using the modified DEPT sequence depicted in Fig. 1b. a, natural abundance spectrum acquired for 3.5 h (5120 scans, TR 2.5s); b, spectrum recorded for 1.8 h (2560 scans, TR 2.5s) during an infusion of 70%-enriched $[1,6-^{13}\text{C}_2]\text{glucose}$. Acquisition was started 1.8 h after the beginning of glucose infusion; c, expansion of b showing the detection of doubly labeled isotomers of glutamate, glutamine and aspartate. Note the triplet of glutamate (Glu-C3T) at 27.9 ppm, corresponding to glutamate labeled simultaneously at the C2, C3 and C4 positions. Processing consisted of zero-filling, filtering (5 Hz lorentzian line broadening for a and 2 Hz lorentzian-to-gaussian resolution enhancement for b and c), fast Fourier transform. No baseline correction was applied. From [125].

sociated with PET measurement of CMRO_2 have also led to skepticism about the validity of the data that generated this controversial concept. The spectroscopic data support the concept that during increased neuronal activity, the fractional increase in CBF significantly exceeds the fractional increase in CMRO_2 and by implication CMR_{glu} which appears to track the changes in CBF.

5. Conclusion

Very high magnetic fields (4 Tesla and above) have been used in human biomedical research only recently and in only few groups. The results so far suggest that while indeed technical challenges increase substantially, there are also significant gains that can be realized in many applications ranging from proton imaging to spectroscopy.

Acknowledgments

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