NRC Publications Archive Archives des publications du CNRC

Implementation of three dimensional wavelet encoding spectroscopic imaging: in vivo application and method comparison

Young, Richard; Serrai, Hacène

This publication could be one of several versions: author's original, accepted manuscript or the publisher's version. / La version de cette publication peut être l'une des suivantes : la version prépublication de l'auteur, la version acceptée du manuscrit ou la version de l'éditeur.

For the publisher's version, please access the DOI link below. / Pour consulter la version de l'éditeur, utilisez le lien DOI ci-dessous.

Publisher's version / Version de l'éditeur:

<https://doi.org/10.1002/mrm.21756>

Magnetic Resonance in Medicine, 61, 1, pp. 6-15, 2008-12-18

NRC Publications Record / Notice d'Archives des publications de CNRC:

<https://nrc-publications.canada.ca/eng/view/object/?id=0aba0b57-9e5f-4f2c-8bfd-e10aad90218c>

<https://publications-cnrc.canada.ca/fra/voir/objet/?id=0aba0b57-9e5f-4f2c-8bfd-e10aad90218c>

Access and use of this website and the material on it are subject to the Terms and Conditions set forth at

<https://nrc-publications.canada.ca/eng/copyright>

READ THESE TERMS AND CONDITIONS CAREFULLY BEFORE USING THIS WEBSITE.

L'accès à ce site Web et l'utilisation de son contenu sont assujettis aux conditions présentées dans le site

<https://publications-cnrc.canada.ca/fra/droits>

LISEZ CES CONDITIONS ATTENTIVEMENT AVANT D'UTILISER CE SITE WEB.

Questions? Contact the NRC Publications Archive team at

PublicationsArchive-ArchivesPublications@nrc-cnrc.gc.ca. If you wish to email the authors directly, please see the first page of the publication for their contact information.

Vous avez des questions? Nous pouvons vous aider. Pour communiquer directement avec un auteur, consultez la première page de la revue dans laquelle son article a été publié afin de trouver ses coordonnées. Si vous n'arrivez pas à les repérer, communiquez avec nous à PublicationsArchive-ArchivesPublications@nrc-cnrc.gc.ca.

National Research
Council Canada

Conseil national de
recherches Canada

Canada

Implementation of Three-Dimensional Wavelet Encoding Spectroscopic Imaging: In Vivo Application and Method Comparison

Richard Young and Hacene Serrai*

We have recently proposed a two-dimensional Wavelet Encoding-Spectroscopic Imaging (WE-SI) technique as an alternative to Chemical Shift Imaging (CSI), to reduce acquisition time and crossvoxel contamination in magnetic resonance spectroscopic imaging (MRSI). In this article we describe the extension of the WE-SI technique to three dimensions and its implementation on a clinical 1.5 T General Electric (GE) scanner. Phantom and in vivo studies are carried out to demonstrate the usefulness of this technique for further acquisition time reduction with low voxel contamination. In wavelet encoding, a set of dilated and translated prototype functions called wavelets are used to span a localized space by dividing it into a set of subspaces with predetermined sizes and locations. In spectroscopic imaging, this process is achieved using radiofrequency (RF) pulses with profiles resembling the wavelet shapes. Slice selective excitation and refocusing RF pulses, with single-band and dual-band profiles similar to Haar wavelets, are used in a modified PRESS sequence to acquire 3D WE-SI data. Wavelet dilation and translation are achieved by changing the strength of the localization gradients and frequency shift of the RF pulses, respectively. The desired spatial resolution in each direction sets the corresponding number of dilations (increases in the localization gradients), and consequently, the number of translations (frequency shift) of the Haar wavelets (RF pulses), which are used to collect magnetic resonance (MR) signals from the corresponding subspaces. Data acquisition time is reduced by using the minimum recovery time (TR_{min}), also called effective time, when successive MR signals from adjacent subspaces are collected. Inverse wavelet transform is performed on the acquired data to produce metabolite maps. The proposed WE-SI method is compared in terms of acquisition time, pixel bleed, and signal-to-noise ratio to the CSI technique. The study outcome shows that 3D WE-SI provides accurate results while reducing both acquisition time and voxel contamination. *Magn Reson Med* 61:6–15, 2009. © 2008 Wiley-Liss, Inc.

Key words: discrete wavelet transform; wavelet encoding; spectroscopic imaging; chemical shift imaging

It is well established that by acquiring metabolic information, magnetic resonance spectroscopic imaging (MRSI) provides a unique modality to noninvasively study tissue metabolism in vivo (1–6). This technique may provide early prognostic information, which aids in better understanding of tissue metabolism, differentiating between diseased and normal tis-

sue, improving treatment, and reducing risk to the patient. However, acquiring this metabolic information in several dimensions is time consuming, especially if the classical Chemical Shift Imaging (CSI) (7,8) technique is used.

To reduce acquisition time, several Fourier-based techniques that use modified high-speed imaging sequences have been proposed (9). Although these rapid techniques, like Echo Planar Imaging (EPI) (10) and spiral imaging (11), provide metabolite information with high spatial resolution and good signal-to-noise ratio (SNR), they require complicated reconstruction methods to differentiate between spatial and spectral information, which leads to a spread of signal over both dimensions (12). Recently, a 2D non-Fourier encoding MRSI technique called wavelet encoding-spectroscopic imaging (WE-SI) using a spin-echo sequence has been proposed as an alternative to CSI to reduce acquisition time and pixel bleed (13). Wavelet encoding has been used in magnetic resonance imaging (MRI) to reduce acquisition time and motion artifacts (14,15). The technique is based upon the discrete wavelet transform that uses dilated and translated prototype functions, called wavelets, to perform a linear transformation from a space domain to a wavelet domain (16). This transform achieves a division of an input finite space function to a set of output subspaces with different sizes and locations. The wavelet dilation determines the size of the subspace, and the translation localizes its position. The number of wavelet dilations, which sets the number of translations, is determined by the desired spatial resolution. Similar to Fourier, which performs an inverse Fourier transform on the k-space data to obtain the input spatial function, an inverse wavelet transform is performed on the subspaces (wavelet domain) data to achieve the same task (16). In practice, WE-SI is similar to the Hadamard technique (17) because it utilizes selective radiofrequency (RF) pulses with profiles resembling the shape of wavelets to excite the subspaces and acquire the corresponding magnetic resonance (MR) signals to fill the wavelet domain. The wavelet dilations and translations are achieved by changing the localization gradient strength and by shifting the frequency of the selective RF pulses, respectively. Inverse wavelet transform is achieved on the collected wavelet domain data to obtain metabolite images. Data acquisition time is reduced by shortening the recovery time (TR) to its minimum value (TR_{min}) in the sequential excitations of predetermined adjacent subspaces (18). Although WE-SI is not as fast as the high-speed Fourier-based techniques cited above, it allows for further reduction in acquisition time if extended to three dimensions, and reduces pixel bleed in the resulting metabolite maps.

Institute for Biodiagnostics (IBD), National Research Council, Winnipeg, MB, Canada.

*Correspondence to: Hacene Serrai, Magnetic Resonance Research & Development, Institute for Biodiagnostics, National Research Council, 435 Ellice Avenue, Winnipeg, Manitoba, R3B 1Y6, Canada. E-mail: Hacene.Serrai@cnrc-nrc.gc.ca
Received 18 February 2008; revised 15 May 2008; accepted June 16, 2008.

DOI 10.1002/mrm.21756

Published online in Wiley InterScience (www.interscience.wiley.com).

© 2008 Wiley-Liss, Inc.

In this article, we briefly describe in the theory section, the extension of the WE-SI from two to three dimensions with acquisition time reduction and SNR calculation, and illustrate its implementation on a clinical 1.5 T GE scanner using a modified PRESS sequence. In the Materials and Methods section, we demonstrate the feasibility of 3D WE-SI in acquiring both phantom and in vivo 3D MRSI data. We show in the Results section that, compared to CSI, 3D WE-SI provides accurate quantification results while reducing acquisition time and preserving the spatial information better. As expected, we observe a reduction in SNR with an increase in resolution. In the Conclusion section, we propose solutions to increase the SNR and improve data quality. We also plan to test the WE-SI technique at higher B_0 field to improve its sensitivity and combine it with parallel imaging for further acquisition time reduction.

THEORY AND METHODS

As described previously (13), the WE-SI technique extended here from two to three dimensions, is derived from the discrete wavelet transform (DWT), which uses three couples of wavelet and scaling functions to encode a finite spatial function $f(x,y,z)$ in three dimensions, up to a given spatial resolution J (16). The wavelet function given by $\psi_{j,k}(x) = 2^{-j/2}\psi(2^{-j}x - k)$ and the scaling function, which obeys a similar relation $\varphi_{j,k}(x) = 2^{-j/2}\varphi(2^{-j}x - k)$, possesses two parameters, the dilation or scale parameter, j , belonging to the set $\{j = 0, 1, \dots, J-1\}$, and the associated translation parameter, k , given by $k = 2^j - 1$. Due to their finite and variable support lengths, wavelet functions act on the input function, $f(x,y,z)$ as localized filters and divide it into a set of localized subspaces with different sizes and positions. The size of each localized subspace is determined by dilating the support length of the wavelet, while its position is defined by spatially translating the wavelet function. For each localized subspace a wavelet coefficient is produced:

$$\left\{ \begin{array}{l} d_1(j,k,l,t) = \int_{x,y,z} \psi_{j,k}(x) \cdot \psi_{j,l}(y) \cdot \psi_{j,t}(z) \cdot f(x,y,z) \cdot dx \cdot dy \cdot dz, \\ d_2(j,k,l,t) = \int_{x,y,z} \psi_{j,k}(x) \cdot \psi_{j,l}(y) \cdot \varphi_{j,0}(z) \cdot f(x,y,z) \cdot dx \cdot dy \cdot dz, \\ d_3(j,k,l,t) = \int_{x,y,z} \psi_{j,k}(x) \cdot \varphi_{j,0}(y) \cdot \psi_{j,t}(z) \cdot f(x,y,z) \cdot dx \cdot dy \cdot dz, \\ d_4(j,k,l,t) = \int_{x,y,z} \psi_{j,k}(x) \cdot \varphi_{j,0}(y) \cdot \varphi_{j,0}(z) \cdot f(x,y,z) \cdot dx \cdot dy \cdot dz, \\ d_5(j,k,l,t) = \int_{x,y,z} \varphi_{j,0}(x) \cdot \psi_{j,l}(y) \cdot \psi_{j,t}(z) \cdot f(x,y,z) \cdot dx \cdot dy \cdot dz, \\ d_6(j,k,l,t) = \int_{x,y,z} \varphi_{j,0}(x) \cdot \psi_{j,l}(y) \cdot \varphi_{j,0}(z) \cdot f(x,y,z) \cdot dx \cdot dy \cdot dz, \\ d_7(j,k,l,t) = \int_{x,y,z} \varphi_{j,0}(x) \cdot \varphi_{j,0}(y) \cdot \psi_{j,t}(z) \cdot f(x,y,z) \cdot dx \cdot dy \cdot dz, \\ a(J,0,0,0) = \int_{x,y,z} \varphi_{j,0}(x) \cdot \varphi_{j,0}(y) \cdot \varphi_{j,0}(z) \cdot f(x,y,z) \cdot dx \cdot dy \cdot dz, \end{array} \right. \quad [1]$$

The wavelet coefficients above are used in the inverse wavelet transform to recover the input spatial function:

$$f(x,y,z) = D_{i=1:7}(j,k,l,t) + A(j,0,0,0). \quad [2]$$

where:

$$\left\{ \begin{array}{l} D_1(j,k,l,t) = \sum_{j=1}^J \sum_{k=0}^{N_j-1} \sum_{l=0}^{N_j-1} \sum_{t=0}^{N_j-1} d_1(j,k,l,t) \cdot \psi_{j,k}(x) \cdot \psi_{j,l}(y) \cdot \psi_{j,t}(z) \\ D_2(j,k,l,t) = \sum_{j=1}^J \sum_{k=0}^{N_j-1} \sum_{l=0}^{N_j-1} d_2(j,k,l,t) \cdot \psi_{j,k}(x) \cdot \psi_{j,l}(y) \cdot \varphi_{j,0}(z) \\ D_3(j,k,l,t) = \sum_{j=1}^J \sum_{k=0}^{N_j-1} \sum_{l=0}^{N_j-1} d_3(j,k,l,t) \cdot \psi_{j,k}(x) \cdot \varphi_{j,0}(y) \cdot \psi_{j,t}(z) \\ D_4(j,k,l,t) = \sum_{j=1}^J \sum_{k=0}^{N_j-1} d_4(j,k,l,t) \cdot \psi_{j,k}(x) \cdot \varphi_{j,0}(y) \cdot \varphi_{j,0}(z) \\ D_5(j,k,l,t) = \sum_{j=1}^J \sum_{l=0}^{N_j-1} \sum_{t=0}^{N_j-1} d_5(j,k,l,t) \cdot \varphi_{j,0}(x) \cdot \psi_{j,l}(y) \cdot \psi_{j,t}(z) \\ D_6(j,k,l,t) = \sum_{j=1}^J \sum_{l=0}^{N_j-1} d_6(j,k,l,t) \cdot \varphi_{j,0}(x) \cdot \psi_{j,l}(y) \cdot \varphi_{j,0}(z) \\ D_7(j,k,l,t) = \sum_{j=1}^J \sum_{l=0}^{N_j-1} d_7(j,k,l,t) \cdot \varphi_{j,0}(x) \cdot \varphi_{j,0}(y) \cdot \psi_{j,t}(z) \\ A(j,k,l,t) = a(J,0,0,0) \cdot \varphi_{j,0}(x) \cdot \varphi_{j,0}(y) \cdot \varphi_{j,0}(z) \end{array} \right. \quad [3]$$

The translation parameters k , l , and t are applied along x , y , and z axes, respectively.

To use wavelet encoding in MRSI, the desired number of voxels, N , is associated with the spatial resolution, $J \{N = 2^J\}$, which sets the number of wavelet dilations, j , and consequently the number of translations, k (16). As stated elsewhere (13), selective RF pulses are used as wavelet functions in WE-SI sequence. They are appropriately represented when the profiles of the RF pulses are similar to their shapes, and their support length is equal to the frequency bandwidth (RF_BW) of the RF pulses. We have chosen Haar wavelets as prototype functions due to their simplicity in implementation. The shape of the Haar scale function is a boxcar, equivalent to a unity vector, resembling the profile of a single-band selective RF pulse, and the shape of Haar wavelet function is a dual boxcar, which corresponds to the profile of a dual band selective RF pulse given by:

$$\psi(x) = \begin{cases} 1 & \text{if } 0 \leq x < 0.5 \\ -1 & \text{if } 0.5 \leq x < 1 \\ 0 & \text{otherwise} \end{cases} \quad [4]$$

Note that the Haar wavelet functions have a finite support in the interval $[k/2^j, (k+1)/2^j]$. Haar functions are used in all directions when encoding the space with WE-SI. Because it achieves a volume selection, PRESS is the appropriate sequence for wavelet encoding in three dimensions. One single-band and one dual-band RF pulse for the excitation and the refocusing RF pulses are generated and inserted in the modified PRESS sequence (19). Translation and dilation

FIG. 1. The modified PRESS sequence and WE-SI algorithm: the wavelet translation and dilation parameters are calculated at each wavelet encoding step to set the sequence parameters: delay time, RF pulse frequency shift, and gradients values. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

operations of the RF pulses are achieved similarly to the 2D WE-SI sequence (13). The slice thickness and the field of view (FOV) determine the gradient strength values in the three spatial directions. The number of voxels, N , in each direction sets the corresponding number of dilations (scale) J , and consequently, the number of translation steps N_j ($N_j = 2^{J-j}$) for each scale value j . The RF pulse is dilated in each direction by multiplying the corresponding gradient strength value by the factor 2^{J-j} , and translated by shifting its central frequency N_j times according to $[(-N_j + 2k + 1)/2]$ RF_BW, where k runs from 0 to $(N_j - 1)$.

The MR signals corresponding to the wavelet coefficients of Eq. [1] are acquired from the subspaces and used in Eq. [2] to recover the image information (13).

Figure 1 displays the modified PRESS sequence used for a 3D WE-SI technique with the wavelet parameters used to set the corresponding gradient and RF frequency shift values prior to RF pulse excitations. Acquisition time is reduced by shortening the TR time to TR_{min} following a frequency shift of the RF pulses (wavelet translation).

Timing in Wavelet Encoding

Due to the finite support of the wavelet functions, the RF pulses cover a part of the space (subspace) for each excitation. Thus, if spins are experiencing the RF pulse excitation in a given subspace, those in the rest of the space are relaxed. By arranging the excitations in an optimal order, a

series of these subspace signals can be acquired without the need of a full TR time (Fig. 2).

Three different types of RF pulse combinations can be used in the 3D WE-SI depending on the desired spatial resolution (13):

1. The first eight wavelet encoding steps require eight TR times because the whole localized 3D space is covered and all spins are excited (Fig. 2A). These eight steps are always executed and set the minimum resolution to two in each direction.
2. The dual band excitation RF pulse is dilated and translated, dividing the corresponding direction into adjacent subspaces (Fig. 2B,C,D). MR signals from these adjacent subspaces can be acquired using TR_{min} . The 3D WE-SI acquisition time is: $Acq_time = (N^2) \cdot (\log_2(N) + 1) \cdot TR$, where N is the number of voxels.
3. The acquisition time in Figure 2B and C can be further reduced by interchanging the excitation RF pulse covering the whole space in a given direction (x, for example) with the dual band refocusing RF pulse covering a subspace in another direction (y, for example) (13). The acquisition time resumes to:

$$Acq_time = N_x \cdot N_y \cdot N_z \cdot TR - N_{eff} \cdot (TR - TR_{min}), \quad [5]$$

where N_x, N_y, N_z are the desired spatial resolution in x, y, and z, respectively, and N_{eff} given by:

FIG. 2. Case of a $4 \times 4 \times 4$ WE-SI sequence: the excitation RF pulse profiles (single and dual band) are colored in black and applied along the z axis, the first refocusing RF pulse is colored in red and applied along the x axis; the second refocusing RF pulse is colored in blue and applied along the y axis. For each wavelet encoding step the corresponding excited region is colored in black. (A) Eight encoding steps are required with no RF pulse translations using eight TR times because the whole space is covered. (B) Twenty-four encoding steps are acquired in 12 TR and 12 TR_{min} with one RF pulse translation. In the upper slab the second refocusing RF pulse is dilated and translated and interchanged with the excitation RF pulse to allow acquisition time reduction. In the middle slab, the excitation RF pulse is dilated and translated with no RF pulse interchange. In the lower slab, the first refocusing RF pulse is dilated and translated and interchanged with the excitation RF pulse. (C) Twenty-four encoding steps are acquired in 12 TR and 12 TR_{min} with two RF pulses translated. In the upper slab both refocusing RF pulses are dilated and translated with RF pulse interchange between the excitation RF pulse and the second refocusing RF pulse. In the middle slab, the excitation and the second refocusing RF pulses are dilated and translated. In the lower slab, the excitation and the first refocusing RF pulses are dilated and translated. (D) Eight encoding steps are acquired in four TR and four TR_{min}. All the RF pulses are dilated and translated with no RF pulse interchange.

$$N_{eff} = \left[\begin{aligned} &2^{2M_x} \cdot (M_z - 2^{M_z} + 6)/3 + 2^{2M_y+1} \cdot (M_z - 2^{M_z} + 5)/3 + 2^{M_x+M_y+M_z+1} - 2 \cdot 2^{M_x} \\ &+ 2^{M_y+M_x} (M_y - 2 \cdot M_z - 8 + M_x) + 2^{M_y} (-2 \cdot M_y + 2 + 2 \cdot M_x) - 4/3 - P(x) - P(y) \end{aligned} \right] \quad [6]$$

is the number of times where the sequence is being executed using TR_{min}.

The following variables are set to: $M_x = \log_2(N_x)$, $M_y = \log_2(N_y)$, $M_z = \log_2(N_z)$, $P(x) = i \cdot 2^{2(i-1)}$, and $P(y) = 2m \cdot j \cdot 2^{2(j-1)}$, where i and j run from 2 to M_x , and M_y , respectively. If $M_x = M_y = M_z = M$, N_{eff} is simplified to:

$$N_{eff} = [2^{3M} + 2^{2M}(3M - 8)/3 - 4/3 - 3P] = [N^3 + N^2(3 \cdot \log(N) - 8)/3 - 4/3 - 3P]. \quad [7]$$

Signal to Noise in Wavelet Encoding

Similar to Fourier transform, which preserves the total energy of the signal for each frequency band, wavelet transform performs the same task for each scale or wavelet dilation (16). However, a decrease of SNR occurs in WE-SI at higher resolutions. This is confirmed by the reduction of the size of the excited subspaces (Fig. 2B,C,D) (13,20). As a consequence, the sensitivity in 3D WE-SI compared to Fourier encoding drops by: SNR_{WE-SI}

FIG. 3. Localization images of the $2 \times 2 \times 2$ (A) and $4 \times 4 \times 2$ (B) phantoms with metabolite locations. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

$= SNR_{CSI} \sqrt{\frac{27 \cdot N_x \cdot N_y \cdot N_z}{(4 + 2 \cdot N_x^2 + 2 \cdot N_y^2 + N_z^2) \cdot (N_z^2 + 2)}}$, which is approximated to

$$SNR_{WE-SI} \approx SNR_{CSI} \cdot (3/N)^{3/2} \quad \text{if } N_x = N_y = N_z \quad [8]$$

MATERIALS AND METHODS

The 3D wavelet encoding SI scheme is implemented on a 1.5 Tesla whole-body GE clinical scanner. Using a GE head coil, both 3D WE-SI and CSI methods are used to acquire phantom and human brain data. Two types of excitation and refocusing RF pulses, single band and dual band, are used in the WE-SI sequence. The excitation RF pulses have

a 7-ms duration and a 2.5-kHz bandwidth, whereas the refocusing RF pulses are 7 ms and 1.5 kHz. Two phantoms, small and large, are made from two rectangular plastic holders containing equally spaced 2×2 and 4×4 holes, respectively. The sizes of the phantoms are 60 mm $\times$ 60 mm and 70 mm $\times$ 70 mm, with the diameters of the holes 25 mm and 14 mm for the small and large phantom, respectively. Cylindrical tubes filled with water solutions containing metabolites with known concentrations, are placed in the holes of the plastic holder, dropped in a container filled with pure water, and immersed in a cylinder filled with canola oil (Fig. 3). Two sets of phantom data, $2 \times 2 \times 2$ and $4 \times 4 \times 2$ are acquired using both WE-SI and CSI. The following acquisition parameters are

FIG. 4. Front axial slice of acquired metabolite spectra for the $2 \times 2 \times 2$ WE-SI (A) and CSI (B). [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

FIG. 5. Front axial slice of acquired metabolite spectra for the $4 \times 4 \times 2$ WE-SI (A) and CSI (B).

used: pulse repetition time (TR) = 2.5 s, echo time (TE) = 135 ms, 12 accumulations, 2.5-kHz bandwidth, 2K points, FOV = 60 mm and slice thickness = 30 mm.

Eight healthy volunteers, between 20 and 30 years of age, with signed consent forms obtained are scanned using both 3D WE-SI and CSI. Localization images are acquired first using gradient-echo sequence to determine the region of interest (ROI). Three sets of MRSI data with different resolutions using both WE-SI and CSI are acquired. The acquisition parameters are as follows: for the $4 \times 4 \times 4$, TR = 2 s, TE = 135 ms, two accumulations, and effective time, $TR_{\min}$ approaching 850 ms; for the $8 \times 8 \times 2$, TR = 2 s, TE = 35 ms, four accumulations, and effective time of about 740 ms; for the $8 \times 8 \times 4$, TR = 1.5 s, TE = 35 ms, and four accumulations. The FOV ranges from 74 mm to 80 mm and the slice thickness from 40 to 60 mm for the eight volunteers, resulting in a mean voxel size of 4.7 cc for the $4 \times 4 \times 4$, 2.35 cc for the $8 \times 8 \times 2$, and 1.2 cc, for the $8 \times 8 \times 4$ resolution. The FOV and slice thickness variation is due to the different sizes of the volunteer heads where body weights ranges from 60 kg to 118 kg. Acquisition bandwidth and data size are set to 2.5 kHz bandwidth and 1K points with 400-ms acquisition duration for all the acquired data. Lipid and water signals are suppressed using the standard saturation bands and frequency selective water suppression RF pulses. Data reconstruction is performed using SAGE software provided by GE for the CSI data and an in-house software using Matlab for the WE-SI data. A line broadening of 2.5 Hz is applied on all

acquired data before the quantification is performed using LC model.

RESULTS AND DISCUSSION

Phantom Results

Figure 4A,B displays the front slice spectra from a $2 \times 2 \times 2$ WE-SI and CSI phantom data, respectively. Similarly, Figure 5A,B shows the front slice spectra of the $4 \times 4 \times 2$ phantom data of the WE-SI and CSI, respectively. In terms of data quantification, WE-SI provides accurate results. Table 1 shows the estimated concentration values vs. the expected ones from the $2 \times 2 \times 2$ WE-SI phantom data with insignificant amount of voxel contamination. Voxel contamination is calculated by dividing the amplitude of any metabolite detected in any voxel by its amplitude in the expected voxel. The obtained results for the front slice in the $4 \times 4 \times 2$ WE-SI phantom data show that WE-SI keeps voxel contamination low (Table 2). Pixel bleed in WE-SI is attributed mainly to two reasons: the first reason is the inverse wavelet transformation in Eqs. [2] and [3], which uses the Haar wavelet and scale functions given by Eq. [4] and the unity vector, respectively (13). As shown in Figure 6, the areas of the bands of the RF pulses are not equal to the areas of the corresponding boxcars shown in red. This is mainly due to the large transition band and the edges of the RF pulse profile. To minimize the profile errors, we replaced in Eq. [3] the values of the Haar func-

Table 1

Absolute Estimated Metabolite Concentrations vs. the Expected Ones with Voxel Contamination Calculations in the $2 \times 2 \times 2$ WE-SI Data

Metabolite Voxel	NAA (mM)		Creatine (mM)		Choline (mM)	
	Estimated	Expected	Estimated	Expected	Estimated	Expected
NAA	61.7	62.5	0	0	0	0
Creatine	0	0	49.2	50	0	0
Choline	0	0	0	0	29.1	30
Water	4.2	0	0.45	0	0	0

Table 2(1)

Percentage of Metabolites in Each Voxel in Arbitrary Units (a.u) from Part 1 of the $4 \times 4 \times 2$ Phantom (Fig. 3B)

Metabolite Voxel	Choline	Alanine	lactates	NAA	Creatine	Leucine	Sarcosine
1 Choline	100	0	0	0	0	0	0
2 Water	0	0	0	0	10	0	0
3 Alanine	0	100	0	0	0	0	0
4 Lactates	0	0	100	0	0	17	0
5 NAA	0	0	0	100	18	0	0
6 Creatine	0	0	0	2	100	0	0
7 Leucine	0	0	10	0	0	100	0
8 Sarcosine	0	0	0	0	0	0	100

Table 2(2)

Percentage of Metabolites in Each Voxel in Arbitrary Units (a.u) from Part 2 of the $4 \times 4 \times 2$ Phantom (Fig. 3B)

Metabolite Voxel	Choline	Sarcosine	Alanine	Creatine	NAA	Lactates	Leucine
9 Choline	100	0	0	0	0	0	0
10 Water	0	0	0	3	0	0	0
11 Sarcosine	0	100	0	0	0	0	0
12 Alanine	0	15	100	0	0	0	4
13 Creatine	0	0	0	100	12	0	0
14 NAA	10	0	0	0	100	0	0
15 Lactates	0	0	0	0	0	100	0
16 Leucine	0	14	0	0	0	0	100

a **b** **c** **d**FIG. 6. Profiles of the RF pulses (blue) used as Haar functions (red) in the WE-SI sequence. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

FIG. 7. In vivo spectra from the second axial slice of the $4 \times 4 \times 4$ WE-SI (A) and CSI (B) data on the top of the gradient echo localization brain image. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

tions (1 and -1) by numbers obtained from the fit of the RF pulse profiles to boxcar shapes (Fig. 6). However, to better solve this problem we plan to use other wavelets with smoother decay and shorter duration (21). The second reason for voxel contamination is the cross-slice phase variation introduced by the frequency shift of the RF pulses. Because wavelet encoding is amplitude modulation, the RF pulses need to be perfectly aligned along with their application axis to produce phased MR signals (13). The frequency shift, introduces a deviation of the RF pulse from its application axis. This results in phase variation, referred to as slice walk in the acquired signal (22). To reduce the effect of this problem, we experimentally calculated the phase variation and manually corrected it, because no automatic phase correction in the operating system of the GE scanner is available (SIGNA LX 9.1). However, these manual corrections are not accurate enough, and are subject to a variety of effects including B_0

and B_1 field variations. With newer versions, the cross-slice phase variation is better handled and automatically corrected. This is demonstrated by the results obtained from the small phantom vs. the large phantom. In the small phantom no RF pulse shift is performed, because the whole space to be imaged is covered. The results show an insignificant amount of voxel contamination (Table 1). In the large phantom, voxel contamination is more pronounced because of the frequency shifts of the RF pulses (Table 2).

Qualitative comparison of CSI results in Figures 4B and 5B show severe pixel bleed when compared to WE-SI. Voxel contamination, which depends on the point spread function (PSF), remains stable in WE-SI, whereas in CSI, it reduces with an increase of the spatial resolution. The PSF in WE-SI is a function of wavelet shape, and consequently, of the RF pulse profile (13,19). The spectral resolution in the acquired data for both WE-SI and CSI data is low, due

FIG. 8. In vivo spectra from the back axial slice of the $8 \times 8 \times 2$ WE-SI (A) and CSI (B) data on the top of the gradient echo localization brain image. [Color figure can be viewed in the online issue, which is available at www.interscience.wiley.com.]

Table 3

The Mean and Standard Deviation Values of the Relative Concentrations of NAA/Creatine and Choline/Creatine Ratios for Different Resolutions

Mean $\pm$ SD	$4 \times 4 \times 4$		$8 \times 8 \times 2$		$8 \times 8 \times 4$	
	NAA/Cre	Cho/Cre	NAA/Cre	Cho/Cre	NAA/Cre	Cho/Cre
CSI	2.05 ± 0.31	0.37 ± 0.11	1.89 ± 0.55	0.31 ± 0.09	1.92 ± 0.47	0.35 ± 0.08
WE-SI	1.77 ± 0.27	0.36 ± 0.03	1.55 ± 0.29	0.27 ± 0.14	1.67 ± 0.25	0.27 ± 0.14

Concentrations from voxels with SNR ≤ 2 in the LC model are removed.

to the difficulties in improving B_0 field homogeneity when using the phantoms (Figure 5A,B).

In Vivo Results

Figure 7 displays the second axial slice spectra from a $4 \times 4 \times 4$ WE-SI and from CSI human subject data, respectively. Figure 8 displays the back axial slice spectra from an $8 \times 8 \times 2$ WE-SI and from CSI human subject data, respectively. Similar relative quantification results of the metabolites are obtained. Table 3 shows the mean values and the standard deviations over all the voxels for the eight volunteers at each spatial resolution of the metabolites ratios of *N*-acetyl-aspartate (NAA) and choline peaks to the creatine peak. The proposed WE-SI technique provides similar results at different spatial resolutions, whereas the CSI method provides higher NAA/creatine ratios at lower resolutions, which tends to decrease at higher resolutions (21). In Table 4, acquisition times are given for both WE-SI and CSI at different resolutions. WE-SI is quicker in acquiring MRSI data than CSI. The experimental acquisition times approach the calculated ones using Eq. [5].

As expected, the SNR is lower in WE-SI than CSI by a factor of 1.5, 2.7, and 3.2 in the $4 \times 4 \times 4$, $8 \times 8 \times 2$, and $8 \times 8 \times 4$, respectively (Table 5). The SNR is calculated as the ratio of the NAA peak intensity and the standard deviation of noise. The SNR values are consistent with Eq. [8]. To increase the SNR in WE-SI, we are testing the so-called independent component analysis (ICA) technique. The ICA uses higher statistical orders, in addition to the mean and the variance, to separate mixed observations collected from independent sources using different cap-tors (24). WE-SI fulfills these conditions because MR signals are acquired from independent subspaces. We will also test other types of wavelets that possess better edge definition and better immunity to noise to compensate for the SNR loss (20,21).

CONCLUSIONS

A 3D wavelet encoding method for acquiring magnetic resonance spectroscopic imaging data is presented. The proposed WE-SI is compared here to the CSI, seen as a reference method, and offering a valuable indication of acquisition time, pixel bleed, and sensitivity. Previous results (13,18) along with the results shown here demonstrate that compared to the CSI method, wavelet encoding technique is able to reduce acquisition time while preserving the spatial distribution of metabolites. The reduction in acquisition time is directly proportional to the spatial resolution and dimensions. Acquisition time could be reduced with CSI using $TR_{\min}$. However, this causes spin saturation due to the long T_1 of the metabolites (1–2 s for protons and 4–6 s for phosphorus), which bias the quantification results. The WE-SI method also collects data with high spectral resolution using just four RF pulses for space encoding, whereas the Hadamard SI technique necessitates a different RF pulse for each row of the Hadamard matrix.

Wavelet encoding seems to suffer from low SNR compared to CSI. To increase the SNR, less spatially localized wavelets should be used (21).

To reduce data reconstruction artifacts, which are the main sources of pixel bleed, crossvoxel phase variation will be addressed in newer operating systems and a smoother wavelet basis that is less dependent on the excitation RF profile shapes should be tested. As a consequence, shorter RF pulses could be used and data with shorter echo times could be acquired.

To further reduce acquisition time, wavelet encoding-SI technique is being combined with the parallel imaging approach (25–28).

ACKNOWLEDGMENTS

This work is sponsored by NSERC-Discovery Grant NSERC RGPIN 327593-06. The authors thank Dr. Lotfi Senhadji of

Table 4

Calculated and Experimental Acquisition Time Duration for Both WE-SI and CSI for Three Different Resolutions

Acquisition Time	$4 \times 4 \times 4$ Experiment	$4 \times 4 \times 4$ Calculated	$8 \times 8 \times 2$ Experiment	$8 \times 8 \times 2$ Calculated	$8 \times 8 \times 4$ Experiment	$8 \times 8 \times 4$ Calculated
CSI	256 s	—	1024 s	—	1536 s	—
WE-SI	186 s	182 s	556 s	560 s*	936 s	940 s

See text for sequence acquisition parameter values.

*The calculated time for the $8 \times 8 \times 2$ is calculated from a 2D equation because the acceleration factor is activated in just two dimensions.

Table 5
Experimental and Calculated SNR Values for the WE-SI and CSI

SNR (a.u)	$4 \times 4 \times 4$	$8 \times 8 \times 2$	$8 \times 8 \times 4$
CSI	20.76	37.59	22.09
WE-SI (experiment)	13.43	13.91	6.87
WE-SI (calculated)	13.48	13.66	6.55

the University of Rennes1, France, and Dr. Robert Kraft of the Wake Forest University, Winston-Salem, NC, USA, for their helpful advice.

REFERENCES

- Gruber S, Stadlbauer A, Mlynarik V, Gatterbauer, B, Roessler, K, Moser E. Proton magnetic resonance spectroscopic imaging in brain tumor diagnosis. *Neurosurg Clin North Am* 2005;16:101–114.
- Hetherington HP, Kim JH, Pan JW, Spencer DD. H-1 and P-31 spectroscopic imaging of epilepsy: spectroscopic and histologic correlations. *Epilepsia* 2004;45:17–23.
- McKnight TR. Proton magnetic resonance spectroscopic evaluation of brain tumor metabolism. *Semin Oncol* 2004;31:605–617.
- Dhinsa R, Qayyum A, Coakley FV, Lu Y, Jones KD, Saanson MG, Carroll PR, Hricak H, Kurhanewicz J. Prostate cancer localization with endorectal MR imaging and MR spectroscopic imaging: effect of clinical data on reader accuracy. *Radiology* 2004;230:215–220.
- Graves EE, Nelson SJ, Vigneron DB, Vignerson DB, Verhey, L, McDermot M, Lawson D, Chang S, Prados MD, Dillon WP. Serial proton MR spectroscopic imaging of recurrent malignant gliomas after gamma knife radiosurgery. *AJNR Am J Neuroradiol* 2001;22:613–624.
- Kurhanewicz J, Swanson MG, Wood PJ, Vigneron DB. Magnetic resonance imaging and spectroscopic imaging: improved patient selection and potential for metabolic intermediate endpoints in prostate cancer chemoprevention trials. *Urology* 2001;57:124–128.
- Brown TB, Kincaid BM, Ugurbil K. NMR chemical shift imaging in three dimensions. *Proc Natl Acad Sci* 1982;79:3523–3526.
- Maudsley AA, Hilal K. Spatially resolved high resolution spectroscopy by four dimensional NMR. *J Magn Reson* 1983;51:147–152.
- Pohmann R, vonKienlin M, Haase A. Theoretical evaluation and comparison of fast chemical shift imaging methods. *J Magn Reson* 1997;129:145–160.
- Posse S, Gioacchino T, Risinger O, LeBihan D. High-speed 1H spectroscopic imaging in human brain by echo-planar spatial-spectral encoding. *Magn Reson Med* 1995;37:644–650.
- Adalsteinsson E, Irrazabal P, Topp S, Meyer C, Macovski A, Spielman DM. Volumetric spectroscopic imaging with spiral-based k-space trajectories. *Magn Reson Med* 1998;39:889–898.
- Hanson LG, Schaumburg K, Paulson OB. Reconstruction strategy for echo planar spectroscopy and its application to partially undersampled imaging. *Magn Reson Med* 2000;44:412–417.
- Serrai H, Senhadji L. Acquisition time reduction in magnetic resonance spectroscopic imaging using discrete wavelet encoding. *J Magn Reson* 2005;177:22–30.
- Healy DM, Weaver JB. Two applications of wavelet transforms in magnetic resonance imaging. *IEEE Trans Info Theory* 1992;38:840–860.
- Panych LP, Jakab PD, Jolesz FA. Implementation of wavelet-encoded MR imaging. *J Magn Reson Imag* 1993;3:649–655.
- Daubechies I. Ten lectures on wavelets. In CBMS-NSF lectures. Philadelphia, PA: Society for Industrial and Applied Mathematics (SIAM); 1992; p SIAM-61.
- Bolinger L, Leigh JS. Hadamard Spectroscopic Imaging (HSI) for multi-volume localization. *J Magn Reson* 1988;80:162–167.
- Serrai H, Young R. Reduction of acquisition time in Magnetic Resonance Spectroscopic Imaging using Wavelet encoding method. *International Society for Magnetic Resonance in Medicine (ISMRM)*, Seattle, May, 2006.
- Cunningham CH, Wood ML. Method for improved multi-band excitation profiles using the Shinnar-Le Roux transform. *Magn Reson Med* 1999;42:577–584.
- Weaver J, Healy D. Signal-to-noise ratios and effective repetition times for wavelet encoding and encoding with wavelet packet bases. *J Magn Reson (A)* 1995;113:1–10.
- Panych LP. Theoretical comparison of Fourier and wavelet encoding in magnetic resonance imaging. *IEEE Trans Med Imaging* 1996;15:141–153.
- Gelman N, Wood ML. Wavelet encoding for 3D gradient-echo MR imaging. *Magn Reson Med* 1996;36:613–619.
- Kreis R, Ernst T, Ross BD. Absolute quantitation of water and metabolites in the human brain. II. Metabolite concentrations. *J Magn Reson (B)* 1993;102:9–19.
- Anemuller J, Duann, J-R, Sejnowski TJ, Makeig S. Spatio-temporal dynamics in fMRI recordings unrevealed using complex ICA. *Lecture Notes Comput Sci* 2004;3195:1103–1110.
- Sodickson DK, Manning WJ. Simultaneous acquisition of spatial harmonics (SMASH): fast imaging with radiofrequency coil arrays. *Magn Reson Med* 1997;38:591–603.
- Lin F, Tsai S, Otazo R, Caprihan A, Wald LL, Belliveau JW, Posse S. Sensitivity-encoded (SENSE) proton echo-planar spectroscopic imaging (PEPSI) in the human brain. *Magn Reson Med* 2007;57:249–257.
- Fu Y, Serrai H. High speed magnetic resonance spectroscopic imaging using wavelet encoding and parallel imaging. *International Society for Magnetic Resonance in Medicine (ISMRM)*, Toronto, May, 2008.
- Pruessmann KP, Weiger M, Scheidegger MB, Boesiger P. SENSE: sensitivity encoding for fast MRI. *Magn Reson Med* 1999;42:952–962.