Rotating-Frame Intermolecular Double-Quantum Spin-Lattice Relaxation $T_{1p,DQC}$-Weighted Magnetic Resonance Imaging

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In this study, spin-locking techniques were added as a part of intermolecular multiple-quantum experiments, thereby introducing the concept of rotating-frame intermolecular double-quantum spin-lattice relaxation, $T_{1p,DQC}$. A novel magnetic resonance imaging methodology based on intermolecular multiple-quantum coherences is demonstrated on a 7.05-T microimaging scanner. The results clearly reveal that the intermolecular double-quantum coherence $T_{1p,DQC}$-weighted imaging technique provides an alternative contrast mechanism to conventional imaging. Magn Reson Med 53:930–936, 2005. © 2005 Wiley-Liss, Inc.

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Spin-lattice relaxation time ($T_{1p}$) in the rotating frame has been demonstrated to be effective in probing slow-motion macromolecules in high static fields (1). The imaging of biologic tissue based on $T_{1p}$ is currently being investigated for various tissues, including articular cartilage (2), the human knee (3), the head, and the neck (4). In $T_{1p}$-weighted MRI of tissues, the image is sensitive to molecular processes that occur over a range of frequencies determined by the amplitude of an applied spin-locking pulse. The image contrast is characterized by the proton relaxation that occurs in the low locking field, but relaxation is unaffected by the much higher Zeeman magnetic field. The dephasing of spins is weaker during the locking pulse than in $T_{1}$-weighted imaging. Additionally, spin-locking imaging is less sensitive to diffusion losses. The $T_{1p}$ MRI provides increased discrimination between normal and diseased tissues because its contrast is a novel early indicator for detecting some diseases (5–9).

Intermolecular multiple-quantum coherence (iMQCs) in highly polarized spin systems (10), which seem to contradict conventional NMR theory and may be crazed in the eyes of NMR spectroscopists because of unexpected cross-peaks in the indirectly detected dimension (10), has led to controversy in the NMR community over the past decade. However, it has gained extensive recognition and is becoming a useful tool in NMR and MRI. Warren and Jeener and colleagues presented a series of studies that characterize the effects of long-range dipolar couplings in iMQCs by classic or quantum treatments, and both approaches have been shown to be equivalent (10–12). Recently, the application of iMQCs to MR imaging has attracted much interest because it has a novel contrast mechanism (13–20). Zhong et al. also successfully obtained meaningful findings in their theoretical research on imaging applications, which concern intermolecular double-quantum coherences (iDQCs) (21–23).

iQCM imaging provides more flexibility in the various parameters for yielding contrast opposed to other imaging methods, such as iDQC relaxation $T_{2,DQC}$-weighted imaging (14,15,17,23). However, the signals obtained using the iDQC $T_{2,DQC}$-weighted method tend to decline very fast with time and are easily perturbed by various harmful artifacts, such as background inhomogeneities and molecular diffusion-induced signal losses (21,23).

This study extends the conventional single-quantum rotating-frame $T_{1p}$ relaxation to intermolecular double-quantum coherences and presents the concept of “intermolecular double-quantum spin-lattice relaxation, $T_{1p,DQC}$, in the rotating frame.” When $T_{1p,DQC}$-weighting is applied to iDQC imaging, the spins are locked in the presence of a spin-locking pulse, minimizing the dephasing of the spins. The diffusion-induced signal losses are substantially lower than those of $T_{2,DQC}$-weighted imaging, because the magnetization is not modulated during the locking period under the experimental conditions used herein. The unique signals are sensitive to the slow motion of macromolecules (1). Furthermore, the $T_{1p,DQC}$ relaxation exhibits “frequency dispersion.” The duration and the strength of the locking field are two further characteristics that can be manipulated to yield the image contrast for various tissues. To our knowledge, rotating-frame spin-lattice relaxation has not yet been investigated in relation to iMQCs. Hence, such an imaging experiment is performed herein.

The effectiveness of $T_{1p,DQC}$-weighted iDQC MR imaging is demonstrated using a 7.05-T microimaging scanner. The results were quantitatively compared with those obtained by conventional $T_{1p}$-weighted SQC imaging. The characteristics and potential applications of this method are clearly analyzed.

THEORY

Signals from iMQCs can be easily observed in simple CRAZED experiments (10). If a field gradient breaks the symmetry of a sample, then higher-order coherences in the thermal equilibrium density matrix may be generated.
through the residual dipolar interaction (11,12,24). These phenomena have been theoretically described using both classic dipolar-field theory and quantum mechanical density matrix treatment (12,24).

Figure 1 displays an imaging sequence based on the CRAZED sequence. The sequence has the following characteristics. (a) A pair of separated correlation gradients with an area ratio of 1:n is utilized to select the desired n-quantum coherences, and the orientation of the correlation gradients was manipulated by altering the angle \( \theta \). (b) The spin-locking pulse labeled \( T_{\text{SL}} \) is applied immediately following the first 90° RF pulse, and the iMQC signals is then modulated by the rotating-frame spin-lattice relaxation. (c) The soft 180° pulse (a “sinc” function) not only acts as a slice-selection pulse, but also removes the inhomogeneity of the magnetic fields. Crusher gradients \( G_6' \) are applied to dephase the signals, which are not the same as those from the selected slice.

Without the high-temperature approximation (10,24), the initial hard 90°-\( \pm \phi \) RF pulse transforms the quadratic terms in the equilibrium density operators into the \( I_jZ_I k \) of the spins \( j \) and \( k \), which involves intermolecular double-quantum and zero-quantum coherences (iZQCs) (10). The product operators of iDQCs and iZQCs can be expressed in terms of the raising and lower operators as

\[
\sigma (0^+) = I_jZ_k
\]

\[
= \frac{1}{4} \{ (I_jZ_k) + (I_jZ_k) + (I_jZ_k) + (I_jZ_k) \}.
\]  

Immediately thereafter, a low-power spin-locking pulse phase shifted by 90° relative to the first pulse was applied to two distant spins. Both were locked in the transverse plane for the duration \( T_{\text{SL}} \):

\[
\sigma (T_{\text{SL}}) = \frac{1}{4} \{ (I_jZ_k) + (I_jZ_k) + (I_jZ_k) + (I_jZ_k) + (I_jZ_k) \} e^{-T_{\text{SL}}/T_{\text{iDQC}}}.
\]

\[
+ (I_jZ_k) e^{-T_{\text{SL}}/T_{\text{iZQC}}}.
\]  

where \( T_{\text{iDQC}} \) and \( T_{\text{iZQC}} \) are defined in terms of the rotating-frame intermolecular two-spin, double-quantum, and zero-quantum spin-lattice relaxations, respectively. Equation [2] indicates that a locking field may simultaneously be applied to substantial separation spins, and each component relaxes with a specific time constant. The cross relaxation between the two spins is very weak, since the considerable distance between them makes their correlation couplings effectively negligible in comparison to the locking field. The Appendix offers a detailed treatment of the intermolecular two-spin term \( I_jZ_k \) during the locking period, using reducible tensor operators.
\[ \sigma(t_s) = \frac{1}{4} (I_x I_x e^{i \gamma H_0 \Delta} + I_y I_y e^{i \gamma H_0 \Delta}) e^{-T_{1,DQC}/T_{1,DQC}} e^{-T_{2,DQC}/T_{2,DQC}}, \]  

where \( t_s \) is the detection time immediately following the \( \beta \) pulse. Finally, the detectable signals are obtained from the \( I_x I_x \) or \( I_y I_y \) terms by removing the \( I_z \) parts in Eq. [4] via the small residual dipole–dipole couplings \( D_{jk} \) (10,24). Therefore, the complex magnetization that originates from iDQCs is given by (24)

\[ M_{iDQC}(t_s) = i M_0 \left( \frac{2\pi}{T_2} \right) I_z(-t_s/\Delta_2) e^{-T_{1,DQC}/T_{1,DQC}} e^{-T_{2,DQC}/T_{2,DQC}} e^{-t_s/T_2}. \]  

Here, \( f_2 \) is the second-order Bessel function; \( M_0 \) is the equilibrium magnetization per unit volume; \( \Delta_2 = \frac{3(S \cdot \hat{Z})^2 - 1/2}{2} \), where \( S \) is the direction of the correlation gradient, \( \hat{G}_0 \), and \( \hat{Z} \) is the static magnetic field direction; \( T_2 \) is the conventional SQC relaxation; \( \tau_{DQC} = (\gamma_j M_0 \mu_0)^{-1} \) is the dipolar field constant, and \( \mu_0 \) is the magnetic permeability constant (24). The observable signals are then proportional to

\[ f_1(T_{SL}, T_{1p,DQC}) \text{ and } f_2(T, T_{1p,DQC}) \text{ represent functions of the relaxations of } T_{1p,DQC} \text{ and } T_{2,DQC}, \text{ respectively. In the iDQC case, both spins } j \text{ and } k \text{ on individual molecules are simultaneously locked in a parallel configuration by the locking field and exponentially decay with a constant } 1/T_{1p,DQC} / T_{1p,DQC} = 1/T_{1p,j} + 1/T_{1p,k}. \text{ For two like spins, the cross relaxation between the pairs of spins is negligible, so the spins have the same relaxation time constant (namely } T_{1p,j} = T_{1p,k} = T_{1p,j} \text{). Accordingly, the following relationship holds; } T_{1p,DQC} = T_{1p,SQC}/2, \text{ where } T_{1p,SQC} \text{ is the time constant of the conventional rotating-frame SQC spin-lattice relaxation. The fact that } T_{1p,DQC} \text{ is only half of } T_{1p,SQC} \text{ may be exploited to detect such ultralow motion molecules without the use of a much longer locking pulse than is used in conventional } T_{1p,DQC} \text{ weighted experiments. Furthermore, } T_{1p,DQC} \text{ weighted iDQC imaging is detrimentally affected by various artifacts, including background inhomogeneities and strong diffusion attenuation. In } T_{1p,DQC} \text{ weighted iDQC imaging, however, the dephasing of locked spins is minimized during the locking period, so the diffusion-induced signal losses are considerably reduced.}

In the conventional \( T_{1p,DQC} \)-weighted imaging sequence depicted in Fig. 2, the initial magnetization is nutated into the \( y \)-axis by the first hard 90° pulse. Then, the magnetization is locked along the \( x \)-axis and decays with an SQC \( T_{1p} \) relaxation rate. Finally, the second hard 90° pulse restores the magnetization to the longitudinal axis. The initial three-pulse cluster appends as a preparatory pulse to the beginning of the standard spin-echo (SE) imaging sequence to yield \( T_{1p,DQC} \)-weighted image contrast. A strong crusher gradient is applied immediately following the preparatory pulse to eliminate any residual transversal magnetization (9). During the locking period, \( T_{SL} \), the signals decay with a time constant \( T_{1p} \) according to

\[ S\left( T_{SL} \right) = S_0 e^{-T_{SL}/T_{1p}}, \]  

where \( S_0 \) is the signal intensity at the shortest \( T_{SL} \).

### MATERIALS AND METHODS

All experiments were performed on a 7.05-T Varian Unity Inova NMR spectrometer (Varian, CA) with microimaging capability. The images were obtained using a microimaging probehead (Resonance Research, Inc., Millarica, MA), which comprises a quadrature birdcage imaging RF coil (30 mm I.D.) and a self-shielded gradient system with a maximum strength of 100 gauss/cm in each of the \( x \)-, \( y \)-, and \( z \)-directions. The pig-tail samples were initially treated with formalin solution and then measured at room temperature (22 to 24°C).

The typical imaging parameters used in the \( T_{1p,DQC} \)-weighted iDQC images, and the conventional \( T_{1p} \)-weighted SE images were field of view, 45 × 30 mm; matrix size, 128 × 128; thickness of the slice, 3 mm; acquisition bandwidth, approximately 50 KHz; spin-locking frequency, in the range 0.77 ~ 1.92 KHz; and echo time, 13 ms. A long repetition time (TR) of 6 s was employed to guarantee that the magnetization was fully recovered, to minimize any possible stimulated echoes, and to eliminate any \( T_2 \) relaxation-dependent contrast. The durations of the hard 90° pulse, the \( \beta \)-pulse, and the slice-selection pulse were 21 \( \mu s \), 28 \( \mu s \), and 2 ms, respectively. In the experiments in this work, we used the pulse...
results and discussion

In the usual CRAZED sequence with two RF pulses separated by the interval $\tau$ plus a pair of fixed correlation gradients, the $n$th order time-domain echoes occur at the delay $n\tau$ following the $\beta$-pulse (27). In the pulse sequence in Fig. 1a, however, the second echo does not appear in the double interval between the two RF pulses, as is typically found when the locking pulse is in effect. In Fig. 2a and b, the second echoes were obtained with spin-locking pulses at 70 Hz and 0.77 KHz, respectively, with different strengths. When the locking field is too weak to fully lock the individual spins, typical spin relaxation occurs and the second echo will arise at $2(T_{SL} + \tau)$, namely 36 ms, as observed in Fig. 2a, in which $T_{SL} = 12$ ms and $\tau = 6$ ms. However, when the locking field is sufficiently strong to lock the observed spins, additional echoes appear $n$ times at $\tau$ from the chosen $n$-quantum. In the case shown in Fig. 2b, the second echoes appear at $2\tau = 12$ ms.

Figure 3a–c presents a series of slice profiles obtained at the locking frequency 0.77 KHz with a duration of 2 ms. In the figures, the correlation gradients $G_0$ were applied along the Zeeman field ($B_0$), at the magic angle (54.7°) with respect to $B_0$, and perpendicular to $B_0$, respectively. Sagittal imaging of this sample was undertaken by considering the slice gradient in the $x$ direction and the readout gradient in the $z$ direction. In principle, the slice profile should be nearly square over a uniform sample when a sinc refocusing pulse is applied. However, the proton density of the pigtail used in our experiments is not uniform; accordingly, the profile is not very sharp. The iMQC theory indicates that intermolecular dipolar interactions will vanish when the correlation gradients are applied at the magic angle, as they are in Fig. 3b. When the correlation gradients are applied perpendicular to $B_0$, the detected signal is only half of the signal that is expected to be obtained along $B_0$ as expected from Fig. 3a and b (10,21,23). The consistency with the theoretical predictions demonstrates that all of the detected signals originated from intermolecular two-spin dipolar interactions, and not in the leakage of SQCs or any residual contamination from other coherences. Furthermore, the slice profiles show that, since the SNR of the DQC signals is low, the phase cycling scheme is critical to ensure the cancellation of single-quantum signals, and long-time signal averaging is applied to improve its SNR (27).

In Fig. 4a–h, the intermolecular $T_{1p,DQC}$-weighted DQC and conventional $T_{1p}$-weighted SE images were both obtained with a locking field of 1.92 KHz for various durations $T_{SL}$. Here, a small interval of $\tau = 500$ $\mu$s was used to minimize the decay of $T_{2,DQC}$. These images clearly reveal the structure of the pig-tail sample, which is a central bone full of bone marrow, surrounded by muscle, fat, and skin. The articular cartilage is located at the joint between the bones. The $T_{1p,DQC}$-weighted images (Fig. 4a–d) clearly presents regions of muscle, cartilage, and outside skin, whereas other tissues are completely invisible. The bone tissue is a porous medium with a high degree of anisotropy; the unmatched correlation gradient will barely influence the bone image. Additionally, the difference between the susceptibility of the bone and that of the bone marrow generates strong inhomogeneities and aggravates the signal sensitivity. In Fig. 4c and d, the articular cartilage was well highlighted more than the muscle and skin.
because the cartilage decays less than other tissues as the locking time increases. In the conventional $T_1$-weighted image (Fig. 3e–h), fat, muscle, skin, and bone are all visible at various locking times. As the locking time increases, the fat, cartilage, and bone decay more slowly than the muscle and skin.

Regions of interest (ROI) analysis was conducted to compare quantitatively various rectangular ROIs of two types of tissues—cartilage and muscle. The backgrounds of the images were all subtracted before the measurements were made. The signal intensity was calculated as the average of the measured values from eight ROIs of each type of tissue. Figure 5a plots the normalized signal intensity versus spin-locking duration $T_{SL}$ at a locking frequency of 0.77 KHz. These curves are fitted by a monoexponential function of $T_{1p,DQC}$ and $T_{SL}$. All iDQC curves clearly reveal that, as the $T_{SL}$ increases from 2 to 50 ms, the intensity of the signal from the cartilage declines to approximately 28% of the original, and that of the muscles falls to about 32%. In the SE case, the signal from the cartilage declines to approximately 49%, and that of the muscles falls to about 52% as the $T_{SL}$ increases from 2 to 50 ms. The signal attenuation is greater for iDQCs than in the case of the conventional SE when the locking time increases. For example, the curves for cartilage show that when the signals decrease to about 30% of their original levels, the corresponding $T_{SL}$ time is around 44 ms for iDQC and 81 ms for SE, which differ by a factor of approximately 2.

Table 1 lists other relaxation values measured at various locking frequencies. Table 1 also reveals that $T_{1p,DQC}$ is about half of the intrinsic $T_1$. Systematic changes in these values are inevitable and may arise from imperfections in gradients and $B_1$ inhomogeneities in experiments. $T_{1p,DQC}$-weighted iDQC imaging of numerous biologic specimens with small SQC $T_2$ values is difficult because it involves rapid transverse relaxation. However, the use $T_{1p,DQC}$ as a weighting parameter in iDQC imaging is feasible because its associated relaxation rate is lower than that of $T_{2,DQC}$.

Figure 5b plots the variation of relaxation time against locking frequency from 0.77 to 1.92 KHz. When the locking field was weak (0.77 KHz), $T_{1p,DQC}$ of cartilage was about 43 ms. However, increasing the locking field (1.92 KHz) increased $T_{1p,DQC}$ by around 35% to 58 ms. These curves reveal that $T_{1p,DQC}$ increases continuously with the locking field strength. This characteristic reveals that the rotating-frame iDQC spin-lattice relaxation, $T_{1p,DQC}$, is responsible for “frequency dispersion,” which is similar to the conventional SQC “$T_1$ dispersion” (3,6,8). The $T_{1p,DQC}$ dispersion reveals that the slow-motion processes are reflected by the appropriate frequency range of the $B_1$ field. When the $B_1$ field is matched to the correlation times of particular biologic tissues, the field can be manipulated to ensure significant dispersion of tissues throughout the specimens.

**CONCLUSIONS**

This work introduces the spin-locking method into CRAZED experiments and extends the work to the rotating-frame intermolecular double-quantum relaxation. A novel MRI technique based on the intermolecular $T_{1p,DQC}$-weighted DQCs for soft tissues is elucidated. The applications of $T_{1p,DQC}$-weighted MR imaging have several advantages, including the following in particular. (1) Signals arise from the intermolecular residual dipolar interaction among distant spins. These signals are sensitive to the heterogeneity of the sample, including the intrinsic heterogeneity in the tissues of the brain (CSF, gray matter, and white matter), over a particular length scale that is set by modifying the correlation distance. (2) $T_{1p,DQC}$ has been shown to be useful for probing slow-motion macromolecules when the $B_1$ field is matched with the correlation times of biologic tissues. (3) The $T_{1p,DQC}$ process has a slower relaxation than the $T_{2,DQC}$ process. However, the method requires a long acquisition time for the following reasons. (1) A long TR with a typical value of 5 s in the
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![Graph](image)

**FIG. 5.** (a) Plot of normalized signal intensities, determined from the ROI analysis presented in Fig. 4a, of the $T_{1\text{p},\text{DQC}}$-weighted iDQC images and the conventional $T_{1\text{p}}$-weighted SE images, against spin-locking durations at a fixed locking frequency of 0.77 KHz. (b) Relaxation times calculated by exponential fitting to the $T_{1\text{p},\text{DQC}}$-weighted iDQC data and the conventional $T_{1\text{p}}$-weighted SE images against the spin-locking frequencies of muscle and cartilage.

Experiments herein is required to recover the $T_1$-relaxing magnetization; (2) phase cycling is critical for obtaining pure MQC, and (3) iDQC signals are much weaker than conventional SQCs. The typical acquisition time in the experiments herein was 5 hr to obtain acceptable signal levels. This time is likely to restrict their typical applications at present. Therefore, the experimental procedure should be improved to increase the efficiency of the acquisition. The main advantage of the $T_{1\text{p}}$ technique is that it can be used to probe the slowly motions of macromolecules. Despite relatively poor sensitivity, the data herein reveal that $T_{1\text{p},\text{DQC}}$-weighted imaging provides information that differs from that obtained using conventional $T_{1\text{p}}$-weighted SE imaging. The method yields an alternative contrast parameter, $T_{1\text{p},\text{DQC}}$, as a useful value in analyzing pathologic changes.

**APPENDIX**

The formulations listed below refer to the Larmor frequency rotating frame. The relevant irreducible tensor operators (29) that are associated with the two homonuclear spin systems of $I = 1/2$ are listed below.

\[
T_{2(1)} = \frac{1}{\sqrt{6}} [3I_1I_2 - I(I + 1)] \tag{A1}
\]

\[
T_{2\text{a}(s)} = -\frac{i}{\sqrt{2}} (I_1I_2^* + I_1I_2) \tag{A2}
\]

\[
T_{2\text{a}(a)} = \frac{1}{\sqrt{2}} (I_1I_2^* + I_1I_2) \tag{A3}
\]

\[
T_{2\text{a}(a)} = -\frac{i}{\sqrt{2}} (I_1I_2^* + I_1I_2) \tag{A4}
\]

In the proposed pulse sequence (Fig. 1a), following the first 90° pulse along the x-axis, the quadratic terms in the equilibrium density matrix $I_1I_2^*$, in terms of irreducible tensor operators $T_{2\text{a}(a)}$, are transformed into

\[
T_{20} - \frac{1}{2} T_{20} - \frac{\sqrt{2}}{2} T_{2\text{a}(s)} . \tag{A2}
\]

This involves two-spin intermolecular DQCs and ZQCs, corresponding to $I_1I_2^*$. When a spin-locking RF pulse is exactly on resonance along the $g$ axis, the time evolution of the density operator under Hamiltonian $H = \omega_1 I_1I_2^*$ is specified by a coupled diff-

**Table 1**

<table>
<thead>
<tr>
<th>Tissue</th>
<th>$T_{1\text{p},\text{SQC}}$ (ms)</th>
<th>$T_{1\text{p},\text{DQC}}$ (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Muscle</td>
<td>63.6 ± 4.3</td>
<td>94.6 ± 4.7</td>
</tr>
<tr>
<td>Cartilage</td>
<td>77.0 ± 5.8</td>
<td>43.9 ± 4.1</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Lock frequency (kHz)</th>
<th>$T_1$ (s)</th>
<th>$T_2$ (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.77</td>
<td>66.8 ± 3.2</td>
<td>39.8 ± 5.2</td>
</tr>
<tr>
<td>1.05</td>
<td>79.4 ± 1.5</td>
<td>42.3 ± 6.0</td>
</tr>
<tr>
<td>1.42</td>
<td>83.2 ± 6.7</td>
<td>45.9 ± 7.7</td>
</tr>
<tr>
<td>1.92</td>
<td>1.1 ± 0.2</td>
<td>101.6 ± 8.6</td>
</tr>
</tbody>
</table>

*Measured as the average of several ROIs of two types of tissues. The data are presented as means with standard errors.
the terms of the two-spin double-quantum operator \( T_{23} \) allows: 

\[
\frac{d}{dt} \begin{bmatrix} T_{20} \\ T_{21}(s) \\ T_{22}(s) \end{bmatrix} = \begin{bmatrix} 0 & -i\omega_1 & 0 \\ -i\omega_1 & 0 & -i\omega_3 \\ 0 & -i\omega_3 & 0 \end{bmatrix} \times \begin{bmatrix} T_{20} \\ T_{21}(s) \\ T_{22}(s) \end{bmatrix}. \quad [A3]
\]

Hence, the operators \( T_{20} \) and \( T_{22}(s) \) are exchanged with each other in the spin-locking \( B_1 \) field by the application of the operator \( T_{21}(s) \). The relevant path is represented as 

\[
\frac{T_{21}(s)}{T_{20}} \xrightarrow{\text{spin-locking \( B_1 \) field}} T_{22}(s). \quad [A4]
\]

Consequently, a mixture of zero-quantum \( T_{20} \) and double-quantum operators \( T_{22}(s) \) survives at the end of the spin-locking field and both are modulated by the strength of the \( B_1 \) field. When only the signals from iDQCs are considered, the terms of the two-spin double-quantum \( T_{22}(s) \) are filtered out by the correlation gradient pairs, with an area ratio of 1:2. During the locking period, the observed iDQC term \( T_{22}(s) \) decays with a time constant \( T_{1p,\text{iDQC}} \), according to rotating-frame intermolecular two-spin, double-quantum longitudinal relaxation.

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