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High Magnetic Field Gradient PGSE NMR in the Presence of a Large Polarizing Field

Paul T. Callaghan, Michal E. Komlosh, and Magnus Nyden*

*Department of Physics, Massey University, Palmerston North, New Zealand; and *Physical Chemistry I, Center for Chemistry and Chemical Engineering, University of Lund, P.O. Box 124, S-221 00, Lund, Sweden*

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A description is given of pulsed gradient spin echo (PGSE) NMR experiments in which large pulsed magnetic field gradients may be required. The design contraints are discussed and, in particular, the problem of the use of large pulsed magnetic field gradients in conjunction with large polarizing fields is considered. Issues addressed concern probe mechanical assembly, current supply requirements, and pulse shape design. We describe a quadrupolar coil with a gradient amplitude of 1.65 T m^{-1} A^{-1} which has been used successfully up to a maximum gradient of around 40 T m^{-1}. A diffusion coefficent of 7.5 × 10^{-16} m^2 s^{-1} has been measured using this system, the lowest yet achieved by PGSE NMR methods.

INTRODUCTION

The use of pulsed magnetic field gradients to measure molecular self-diffusion was first demonstrated by Stejskal and Tanner in 1965 (*1*). In this technique two identical gradient pulses of amplitude g, duration δ, and separation Δ are applied in the dephasing and rephasing segments of a spin echo or stimulated echo sequence. Since then the pulsed gradient spin echo (PGSE) NMR method has been widely applied and extended to studies of restricted diffusion and flow. One of the more intriguing aspects of this method has been its analysis in terms of a scattering formalism (*2*), where the magnitude of the scattering wave vector q is proportional to the area under the pulses. This q-space approach has been widely used to describe restricted diffusion and flow in terms of the physics of diffraction (*3–8*). Furthermore, PGSE NMR has been used in a number of multidimensional experiments including dynamic NMR imaging (*2*), velocity exchange spectroscopy (*9*), and diffusion ordered spectroscopy (*10*), and recently the use of multiple gradient pulses has been shown to provide access to the spectrum of translational motion in the frequency domain (*11*).

Most PGSE NMR experiments are performed using gradient amplitudes on the order of a few teslas per meter, with a very few researchers reporting work carried out at around 10 T m^{-1} (*12–18*). In most cases, however, the mean squared displacements being measured are quite large ($>10^{-7}$ m) so that sensitivity to artifacts which become apparent at high resolution (around 10^{-8} m) is not obvious. At gradients in excess of 10 T m^{-1} and especially when one wishes to probe the distance scale below 10^{-7} m, a number of problems emerge, some of which are exacerbated when a superconducting magnet is used. In particular, the problem of eddy current induction becomes more significant and the result of the large polarizing field means that Lorentz forces on the wires are consequentially increased. One of us has carried out high spatial resolution experiments over several years using a coil capable of 20 T m^{-1} gradient amplitude which was incorporated in the 60 MHz ^{1}H-NMR probe of an electromagnet (*12*). Recently, we have developed a PGSE system capable of pulsed gradients in excess of this value, using a coil which is incorporated in the 300-MHz probe of a wide-bore superconducting magnet. This paper reports on the use of that apparatus and some of the issues involved in the design of very high magnetic field gradient coils for use in superconducting magnet environments.

There exist a number of experiment classes where very large magnetic field gradients (in excess of 10 T m^{-1}) are important. First, there is the case of very slow diffusion where the mean squared displacements of the spins may be very small over the time scale available to the spin echo (T_2) or the stimulated echo (T_1). A well-known example of such an application concerns the measurement of internal translation modes of a high molecular weight polymer random coil, whose end-to-end distance may be on the order of 1000 Å. In particular, to reliably detect motion on the order of 100 Å, one requires $q > 10^7$ m^{-1} in order to obtain a measurable echo attenuation. Given $q = (2\pi)^{-1}\delta g$ and constraining δ to be no longer than a few tens of milliseconds, we find that we require $g > 10$ T^{-1}. Second, where the spin relaxation is very fast, the need to limit the duration of the gradient pulse to times less than or on the order of T_2 places a severe constraint on the available q amplitude. For example, a limitation to 1 ms means $q^{-1} < (3\ \mu\text{m})^{-1}$ for $g = 10$ T^{-1}. Third, in experiments where there exists a characteristic time scale for molecular motion, for example, the time for diffusion across the pore in the case of restricted diffusion, it is often important to restrict the duration of the gradient pulse in order to satisfy the so-called ''narrow pulse''

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condition (*2*). This condition permits the echo amplitude to be simply described in terms of an ensemble average of phase shifts arising from displacements over the time Δ elapsing between the sharply defined gradient pulses. This temporal restriction again calls upon the availability of large gradient amplitude in order to access as large as possible a value of q. Finally, the use of multiple gradient pulses in the so-called frequency domain modulated gradient spin echo (FG-MGSE) NMR method has a need for very large gradient pulses because of the well-known reduction in sensitivity to translational motion which occurs when magnetic field gradients are inserted between closely spaced 180° RF pulses in a repetitive CPMG train (*11*).

One solution to the problem of how to generate large magnetic field gradients is offered by the stray field method of Kimmich and co-workers (*13*). When a steady gradient is used in combination with a stimulated echo RF pulse sequence, the effect is to cause the effective gradient to be nonzero only during the periods of transverse magnetization evolution (i.e., single quantum coherence) and zero during periods of z-storage (i.e., polarization). By utilizing the stray field in the outer bore of a superconducting magnet it is possible to achieve 20 T m^{-1} without difficulty, and in the case of a specially constructed Maxwell pair superconductor (*14*), around 200 T m^{-1}. However, the stray field method suffers from three principal defects. First, there is a loss of spectral resolution due to the need to acquire the signal in the presence of the gradient. Second, the RF excitation of the spins and subsequent acquisition in the presence of the gradient implies that only a narrow slice of sample will participate and that the signal must be detected at wide spectral width, both effects causing a severe reduction in available signal-to noise ratio. Finally, the inability to control the magnitude of q except by varying the duration of transverse magnetization evolution makes it difficult to separate the influences of relaxation, dipolar dephasing, and gradient dephasing, thus leading to some ambiguity in data interpretation unless special steps are taken to elucidate these effects.

It has been shown recently that very large pulsed magnetic field gradients are possible by utilizing the stray fields in the vicinity of a current-carrying single wire (*19*). Such arrangements are somewhat unusual in that it is possible to place the sample in close proximity to the wire and thus gain advantage of the divergence in gradient strength which occurs in the wire vicinity. In the present article we shall be concerned with the generation of large magnetic field gradient pulses using more conventional current/coil techniques. Before describing our system, it is appropriate to consider the factors which limit the magnitude of gradient which can be achieved using currents pulsed in wires. In particular it may be shown that this limit is determined entirely by the heating of the gradient coil by the current, and, more specifically, the allowed ohmic power density. The scaling laws are as follows. Consider any given coil configuration of N turns whose linear dimensions scale as r. Suppose further that the coil is carrying a current I. The gradient amplitude g is proportional to INr^{-2} while the resistance of the coil, which is both proportional to the length of wire (Nr) and inversely proportional to cross-sectional area (r^2/N), scales as $\rho N^2 r^{-1}$, where ρ is the wire resistivity. Thus, the power dissipated per unit volume of the coil (and hence the associated temperature rise) varies as ρg^2, independent of turn number or coil size. The maximum gradient amplitude is thus limited solely by the permissible power density and the wire resistivity.

In consequence, there is no particular advantage in making gradient coils large or small as regards the power density limit. However, there are other reasons why small coils offer significant advantages, not the least of which are the reduced inductance, the reduced power supply requirements, the reduced Lorentz torque on the coil array, and the reduced induction of eddy currents in the surrounding magnet due to stray pulsed magnetic fields. We note, in particular, that because a gradient coil comprises an opposed pair of dipoles it represents an octopole term and its stray fields therefore attenuate as r^{-5}.

APPARATUS DESCRIPTION

We have designed and built a gradient coil set which is as small as possible, consistent with an interior sample space and RF coil. This coil system is designed to be used in a standard wide (89 mm) bore 7-T magnet and, in particular, to plug directly on the RF tuning stage of a Bruker microimaging probe. However, the design is quite general and could be used with other probe configurations. The schematic layout of the coil is shown in Fig. 1a, and the schematic layout of the wire bundle in relation to the sample space is shown in Fig. 2a. A photograph of the coil assembly is shown in Fig. 1b. It comprises a quadrupolar array with 23 turns per bundle, each bundle being wound between two holes drilled in supporting posts. The 0.36-mm diameter enameled copper wires, while under tension, are initially free standing, but subsequently potted in epoxy (Araldite, Ciba-Geigy) which penetrates the bundle and, once set, provides mechanical support. At the array center a three-turn 4-mm diameter solenoidal RF coil is wound around a horizontal glass supporting tube, thus providing a 3-mm diameter sample space. The 90° pulse time of this coil at 50 W and 300 MHz is 4 μs. The gradient coil has an inductance of 15.3 μH and a resistance of 0.8 Ω.

Figure 2b–d shows the calculated gradient distribution over the 2.0-mm inner diameter of the sample tube space. The uniformity is very good, with a standard deviation of 2.5% and a mean gradient strength of 1.65 T m^{-1} A^{-1}. By comparison, the uniformity is somewhat poorer over a 3-mm diameter as shown in Fig. 3b–d, the standard deviation being 8%.

For PGSE NMR measurements at $q \gtrsim 10^7$ m^{-1} the coil is driven using current pulses with up to 40-A amplitude, thus providing a maximum gradient of 48 T m^{-1}. However, the requirements for stability at such amplitude are very stringent and we have found that several precautions are needed in order

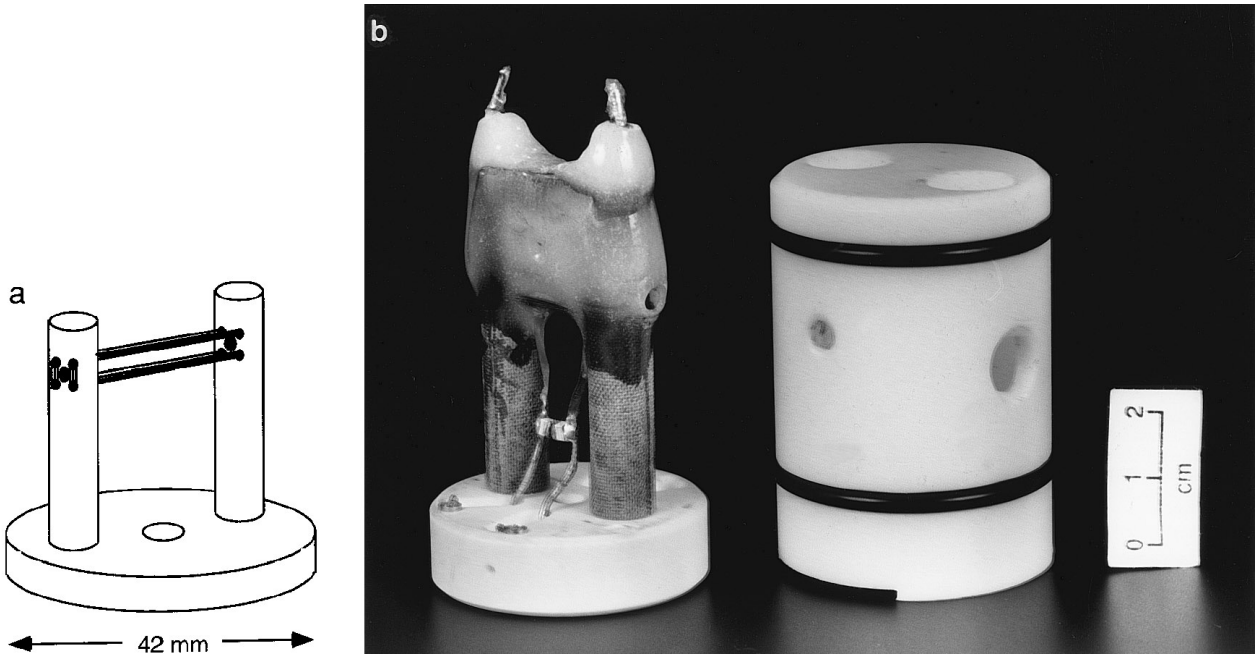

FIG. 1. (a) Schematic diagram of the gradient coil assembly. (b) Photograph of coil assembly and mechanical cowling.

to achieve the desired 100-Å resolution. There is a general maxim concerning PGSE NMR which can be summarized as follows. *All artifacts cause excess echo attenuation, never reduced echo attenuation. All PGSE NMR systems have a lower limit of mean squared displacement below which artifactual attenuation exceeds diffusive attenuation.* In consequence, the diagnostics to be used in testing a PGSE NMR system should be based on the use of a sample for which the mean squared displacements of the spins are as small as possible. We use a very high molecular weight semidilute polymer sample (5% w/v 20×10^6 Da polystyrene in deutero-toluene) in which the segmental mean squared displacements over the PGSE time Δ are exceedingly small ($\sim$100 Å) for Δ on the order of T_2 (around 20 ms). Because the echo attenuation due to diffusive motion is small, attenuation due to spurious effects become dramatically obvious. By this means we are able to progressively improve the system.

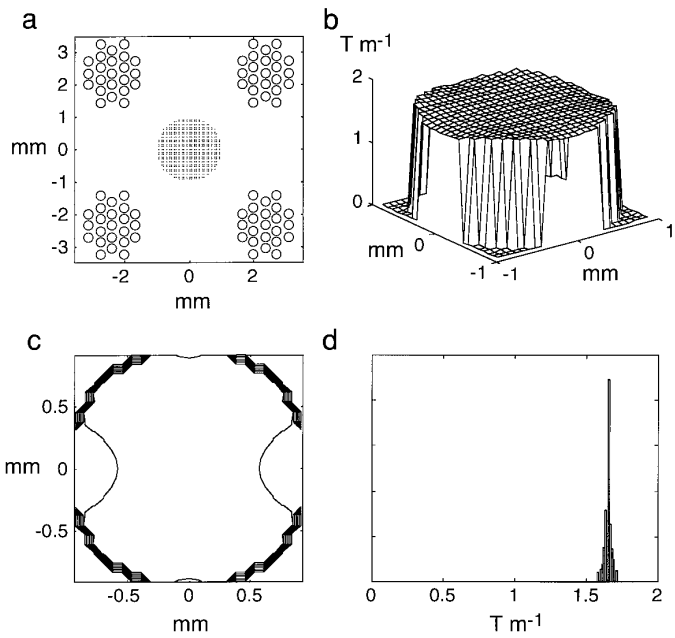

FIG. 2. (a) Layout of the wire bundle in relation to the sample space of 2 mm diameter shown in cross-section. (b) Stackplot profile of gradient across sample space. (c) Contour map of gradient distribution across sample space. (d) Histograph of gradient distribution across 2-mm diameter sample space.

The first problem to be overcome concerned that due to mechanical motion of the coil set, an effect not obvious at 1.4 T but very important in a 7-T environment. To this end we designed a Teflon cowling with locking screws which sits tightly around the gradient coil assembly. The cowling is shown in the photograph in Fig. 1b. The cowling is then held firmly inside the bore of the Bruker microimaging gradient coil assembly by means of two rubber O-rings. The current leads which enter the bore from above are well supported in the vicinity of the coil so that no whiplash motion is transferred. Finally, the sample tube is securely located in the bore of the RF tube by means of Teflon tape. Each of the these steps was found to be crucial to the operation of the system at the required spatial resolution.

The second problem concerned the choice of power supply. All active current sources have noise and hum. At 100-Å resolution the matching required in the gradient pulse areas is on the order of 10 ppm, a level below that normally

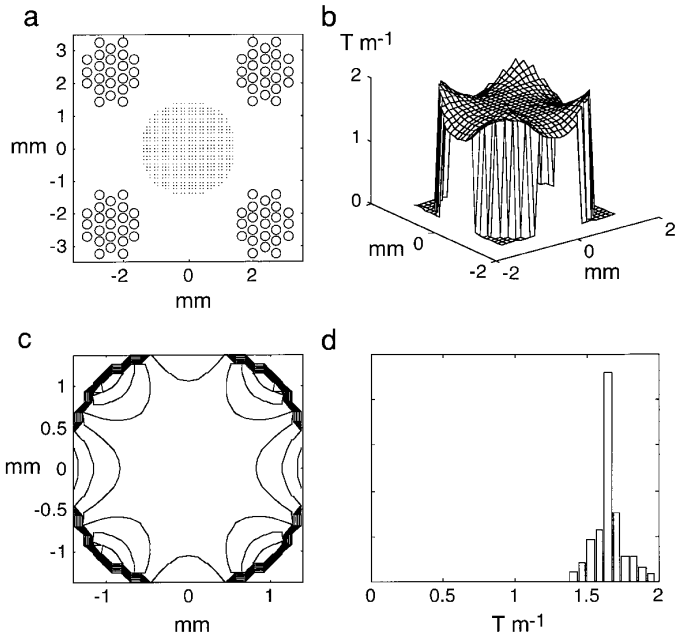

FIG. 3. As for Fig. 2, but for a 3-mm diameter sample space.

measurable by electronic means, but easily detectable in the diagnostic PGSE NMR signal. Of a range of well-known commercial power supplies which we have compared, only the Bruker BAFPA series proved of sufficient quality. These systems are unusual in their use of toroidally wound transformers whose stray mains frequency fields are very small. We have found, however, that these supplies respond best when the current pulse is ramped on the leading and trailing edges using around 20 steps, each of a few microseconds duration, thus giving a controlled rise and fall time on the order of 50 μs. We achieve this by means of the gradient waveform memory control in the Bruker AMX-300 microimaging accessory.

Finally, we note that the use of a few dummy gradient pulses applied just before the PGSE pair significantly improved the performance of the system at the highest spatial resolution limit. We attributed this effect to the settling of the power supply into a steady-state mode of current delivery.

RESULTS

In the narrow gradient pulse limit, the normalized echo amplitude in the PGSE NMR experiment is given by the ensemble average

$$E(q) = \langle \exp(i2\pi qZ) \rangle, \qquad [1]$$

where Z is the displacement of a given spin over the time Δ.

For an ensemble of molecules undergoing unrestricted Fickian self-diffusion, the distribution of mean-squared displacements will be Gaussian and so

$$E(q) = \exp(-4\pi^2 q^2 \langle Z^2 \rangle). \qquad [2]$$

An example of such Gaussian behavior is shown in Fig. 4 for the a 5% solution of 1.57×10^6 dalton polystyrene in perdeuterated toluene. This experiment was performed with Δ = 1000 ms and $g = 14.85$ T m^{-1} and with q varied by changing the pulse duration, δ. The mean squared displacement, $\langle Z^2 \rangle$, is on the order of 5000 Å for this example.

We have been interested in using this gradient coil system to proble smaller mean squared displacements, and in particular the internal modes of higgh polymers. Here the motion is non-Fickian and Eq. [2] cannot be used. In general, for any stochastic displacement distribution, the Taylor expansion for E gives

$$E(q) \approx 1 - 4\pi^2 q^2 \langle Z^2 \rangle + \ldots \qquad [3]$$

so that the semilogarithmic plot ln(E) vs q^2 will be approximately linear down to attenuation values of around 0.8, and yield the mean squared displacement $\langle Z^2 \rangle$ directly.

There are a number of theories concerning the temporal dependence of $\langle Z^2 \rangle$ in the case of random polymers, all of which predict a linear dependence on Δ for $\langle Z^2 \rangle$ in excess of the polymer dimension and a weaker dependence below that. An effective test of the PGSE NMR measurement involves the determination of $\langle Z^2 \rangle$ for polymer segments over a wide time scale for Δ. For the purpose of this article we show data obtained using the semidilute system of 5% w/v 4×10^6 Da polystyrene in deutero-toluene. The polymer was obtained from Polymer Laboratories (Church Stretton, Shropshire, UK) and had a polydispersity index M_W/M_n = 1.06.

Figure 5 shows the dependence of $\langle Z^2 \rangle$ on Δ from Δ = 6 ms to Δ = 3 s. Clear transitional behavior is apparent at $\langle Z^2 \rangle \approx$ 800 Å. We note that this is quite consistent with the

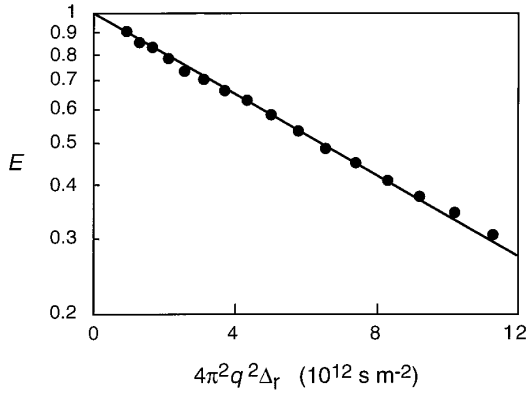

FIG. 4. Echo attenuation plot for a sample of 5% w/v 1.75×10^6 Da polystyrene in *d*-toluene for a diffusion time Δ = 1000 ms. The experiment was performed by keeping g constant at 14.75 T m^{-1} and varying the pulse duration δ.

predicted end-to-end distance of 4×10^6 Da polystyrene under these solution conditions (*20*). The scaling behavior below this transition is consistent with $\langle Z^2 \rangle \sim \Delta^{1/2}$. We are not concerned in this paper with discussing this scaling in detail, but we note that such an exponent would be predicted by most theoretical treatments. The quality of the data is sufficiently good to permit comparative tests of rival theories concerning polymer dynamics. An extensive investigation of $\langle Z^2 \rangle$ vs Δ behavior over a range of polystyrene molecular weight and concentrations will be published elsewhere. For the purpose of the present exercise it is sufficient to show that the data are well behaved and that the spin displacements measured here are reliable. At the shortest duration, Δ, the rms displacement is 270 Å. A center-of-mass diffusion coefficient for the polymer may defined in the long time limit. For this polymer system the value is $(8.0 \pm 0.1) \times 10^{-15}$ m^2 s^{-1}.

As an example of how small a diffusion coefficient may be measured using our apparatus, we shown in Fig. 6 the echo attenuation plot for a 20% solution of 3×10^6 Da polystyrene in the same deutero-toluene solvent. The value obtained is 7.5×10^{-16} m^2 s^{-1}. To our knowledge this value is lower by a factor of 3 than any published value obtained using NMR field gradient methods.

CONCLUSION

While it is relatively straightforward to achieve pulsed gradient amplitudes well in excess of 10 T m^{-1}, the conditions for their effective use involve stringent constraints on pulse area matching, eddy current suppression, and sample movement inhibition. These in turn place constraints on the mechanical mounting to be used, the power supply quality, and the system of pulse generation and rise time control. We have found that a small quadruplar coil array securely mounted in a supporting manifold, driven using a highly stable power supply using ramped current switching and preceding dummy pulses, can give excellent results when used in conjunction with a 7-T superconducting magnet. The results presented here show that resolution on the scale of a few hundred angstroms is possible and that one may certainly measure diffusion coefficents below 10^{-15} m^2 s^{-1}.

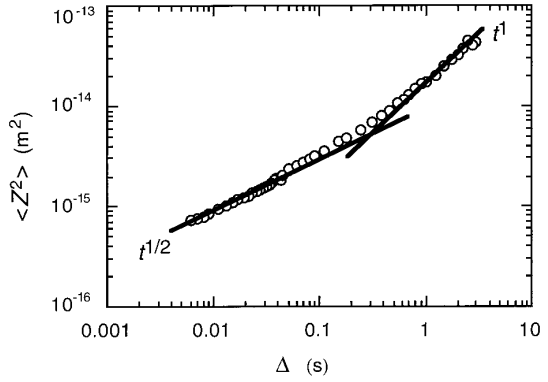

FIG. 5. Mean-squared displacement $\langle Z^2 \rangle$ vs diffusion observation time Δ for a sample of 5% w/v 4.0×10^6 Da polystyrene in *d*-toluene for a diffusion time over the range $\Delta = 6$ ms to $\Delta = 3$ s. Note the transition from Δ^1 to $\Delta^{1/2}$ scaling once the segmental displacements become smaller than the coil dimensions.

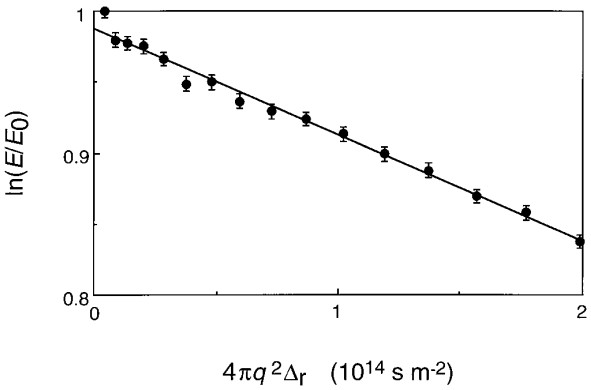

FIG. 6. Echo attenuation plot for a sample of 20% w/v 3.0×10^6 Da polystyrene in *d*-toluene for a diffusion time $\Delta = 1000$ ms. The experiment was performed by keeping g constant at 24.75 T m^{-1} and varying the pulse duration δ. Note that the diffusion coefficient thus obtained is 7.5×10^{-16} m^2 s^{-1}.

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