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A Nonplanimetric Technique for Measuring Fluid Volumes Using MR Imaging—Phantom Results

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Volumetric studies of ventricular and sulcal size have been shown to correlate with normal aging and dementia and may be of use in diagnosing and monitoring Alzheimer's disease [1, 2]. These CSF volume studies make use of planimetric techniques applied to CT images. In this paper we discuss a simple nonplanimetric algorithm for calculating fluid volumes using data from a single-slice MR scan. The algorithm is tested against data from a number of simple phantoms and its possible application to the calculation of intracranial CSF volumes is discussed.

Theory

In MR, assuming uniform sensitivity for signal detection within the imaging field of view (FOV), the signal induced in an MR receiver coil by a uniform fluid volume within the FOV is directly proportional to the density of spins [3] and, hence, to the fluid volume itself. Therefore, doubling the volume of fluid doubles the signal output, all parameters of the pulsing sequence being held constant. A simple ratio of fluid volumes and signal strengths then allows the determination of an unknown fluid volume from a known fluid volume and the measured signal strengths:

\[ VX = \frac{(SX \times VC)}{SC} \]  

(1)

where \( VX \) is the unknown volume, \( SX \) is the signal from this unknown volume, \( VC \) is the known volume of the same type of fluid, and \( SC \) is the signal from this known volume.

To obtain an image of the volume, an FOV and slice thickness that will encompass the total volume are chosen; the result is a projection of the volume onto the image plane. At every point on the image the image intensity is directly proportional to the inverse Fourier transform of the signal [4], and the sum of all pixels in the image is directly proportional to the signal strength (evaluated at the echo center for a spin-echo acquisition) [5]. Equation 1 can thus be written so that the unknown volume can be determined from a summation of image pixels:

\[ VX = \frac{(SUMX \times VC)}{SUMC} \]  

(2)

where \( SUMX \) is the sum of the image pixels for the unknown volume and \( SUMC \) is the sum of the pixels for the known volume. The images of the known and unknown volumes can be obtained from the same acquisition by separating the volumes so that they appear as separate images in the reconstruction.

Materials and Methods

Phantom Construction

A number of phantoms were used to test equation 2. These phantoms, which were filled with distilled water doped with copper sulfate (0.7 g/l of CuSO4·5H2O), are shown in Figure 1. On the left is a flask filled with 500 ml of solution, in the center is a simple brain phantom, and on the right is a plastic bottle filled to the top with 6-mm-diameter glass balls and 100 ml of the doped water. The brain phantom used two sizes of plastic tubing (2.4 mm and 6.4 mm inside diameter) filled with doped water and knotted up to simulate sulci. Embedded in this tangle of tubes were two bottles filled with doped water to simulate ventricles. The total volume of fluid in the brain phantom was 135 ml (85 ml in the tubes; 50 ml in the two bottles). The bottles could be removed to give a phantom without ventricles. All volumes were measured to a 1% accuracy by volumetric or gravimetric means.
Data Acquisition

The data were acquired on a Philips 1.5-T Gyroscan superconducting imager operating at 0.5 T. The known calibration volume (VC in equation 2) was a 5-ml syringe filled with the same doped water as used in the phantoms. Figure 2A shows the brain phantom and syringe positioned on the patient table in preparation for a scan.

One 15-cm-thick transaxial slice image of each phantom with the syringe was obtained with a spin-echo pulse sequence of TR = 250 msec and TE = 50 msec. These particular TR and TE values were not essential to the phantom results. To test this last point the brain phantom data were also acquired with TR = 1000 msec and TE = 30 msec. Real images were acquired with a 300 mm FOV at various resolutions and interpolated into a 256 x 256 matrix for display. The receiver coil was the standard body coil used for clinical imaging. The uniformity of this coil was tested by acquiring an image of a cylinder of fluid contained in the coil. This study indicated a variation of less than 3% in uniformity within the useful FOV.

Figure 2B is the resulting reconstructed image for the brain phantom, along with a rectangular ROI drawn about the phantom image and within which the pixels are summed to obtain the signal from this phantom (SUMX in equation 2). A smaller ROI (not shown) was also drawn about the syringe image in order to obtain its signal (SUMC in equation 2).

Results

The results of the phantom studies are shown in Table 1. The calculated values shown are the average and standard deviation from five separate acquisitions for each case. In all

![Image]

**TABLE 1: Summary of Volume Measurements**

<table>
<thead>
<tr>
<th>Line</th>
<th>Imaged Volume</th>
<th>True</th>
<th>Calculated*</th>
<th>TR (msec)</th>
<th>TE (msec)</th>
<th>Resolution</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Flask</td>
<td>500</td>
<td>506 ± 2</td>
<td>250</td>
<td>50</td>
<td>256x256</td>
</tr>
<tr>
<td>2</td>
<td>Jar with beads</td>
<td>100</td>
<td>99 ± 2</td>
<td>250</td>
<td>50</td>
<td>256x256</td>
</tr>
<tr>
<td>3</td>
<td>Jar with beads</td>
<td>100</td>
<td>99 ± 2</td>
<td>250</td>
<td>50</td>
<td>128x128</td>
</tr>
<tr>
<td>4</td>
<td>Jar with beads</td>
<td>100</td>
<td>98 ± 1</td>
<td>250</td>
<td>50</td>
<td>64x64</td>
</tr>
<tr>
<td>5</td>
<td>Brain with ventricles</td>
<td>135</td>
<td>134 ± 1</td>
<td>250</td>
<td>50</td>
<td>256x256</td>
</tr>
<tr>
<td>6</td>
<td>Brain with ventricles</td>
<td>135</td>
<td>135 ± 1</td>
<td>1000</td>
<td>30</td>
<td>256x256</td>
</tr>
<tr>
<td>7</td>
<td>Brain without ventricles</td>
<td>85</td>
<td>85 ± 4</td>
<td>250</td>
<td>50</td>
<td>256x256</td>
</tr>
<tr>
<td>8</td>
<td>Brain without ventricles</td>
<td>85</td>
<td>88 ± 6</td>
<td>250</td>
<td>50</td>
<td>128x128</td>
</tr>
<tr>
<td>9</td>
<td>Brain without ventricles</td>
<td>85</td>
<td>83 ± 5</td>
<td>250</td>
<td>50</td>
<td>64x64</td>
</tr>
</tbody>
</table>

* Errors shown are the standard deviations.
cases the volumes calculated by equation 2 are within 3% of the true volumes.

The brain phantom data were acquired with two different pulse sequence parameters, and as expected the calculated volumes were not dependent on the sequence used (lines 5 and 6 of Table 1).

The jar with beads and the brain phantom without the ventricles were used to test the dependence of the calculated volume on acquisition resolution. These phantoms were used because of the small sizes of the fluid spaces relative to the imaging resolution (pixel resolutions of 4.7, 2.3, and 1.2 mm for the 64 x 64, 128 x 128, and 256 x 256 acquisitions, respectively). No significant dependence of the calculated volumes on resolution is seen (lines 2, 3, 4 and 7, 8, 9 of Table 1).

The volumes in Table 1 where calculated with real (magnitude and phase data) images, although, in principle, modulus (magnitude only) images may also be used. However, the pixel sums contain a background contribution from noise, which tends to cancel from the real image, whereas for modulus images it results in a small addition to the pixel sums. Edge-ringing artifacts for the low-resolution acquisition were also more of a problem with the modulus images. Volumes calculated with modulus images were approximately 5% below those calculated with the real images.

Discussion

The phantom results in Table 1 indicate that under ideal conditions and in vitro, equation 2 can be used to yield nonplanimetric volume calculations to an accuracy of a few percent over a wide range of phantom volumes. The insensitivity of the volume calculations to the acquisition resolution implies that the scan time could be minimized, for any TR, by decreasing this resolution. This idea could in fact be taken to its limit by acquiring only one line of image data. In this case one would obtain not a two-dimensional image, but a projection; the only requirement would be that the projection be acquired so as to separate the unknown volume’s signal from the known volume’s signal.

From a clinical standpoint, the phantom results imply that intracranial CSF volumes might be determined by comparing signals derived from intracranial CSF with those from known volumes of CSF. For such a procedure to work it would be necessary to use a pulse sequence that would return a signal from the CSF only, such as a spin-echo sequence with very long TE. The small size of the known volume (5 ml) was in fact chosen in anticipation of applying the technique to the clinical situation. This clinical use of the technique will require that a number of questions be addressed, among which are the possible difference between the MR parameters for intracranial and in vitro CSF, and the possible effects from CSF flow, particularly because of the long echo times that would be used to turn off all but the CSF signal.

REFERENCES