Improvement of Phase-Contrast Flow Measurements: Opposite Directional Flow-Encoding Technique to Eliminate the Influence of the Maxwell Term Phase Errors

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Abstract
A new method termed the opposite directional flow-encoding (ODFE) technique is proposed to increase the accuracy and the reproducibility of phase-contrast flow measurements by correcting the non-linear background of velocity images induced by concomitant magnetic fields (Maxwell terms). In this technique, the volume flow rate is calculated from the difference of two region of interest (ROI) values derived from two velocity images obtained by reversing the flow-encoding direction. To evaluate the technique, various phantom experiments were carried out and volume blood flow rates of internal carotid arteries (ICAs) were measured in four volunteers. The technique could measure the volume flow rates of the phantom with higher accuracy (mean absolute percentage error = 1.04%) and reproducibility (coefficient of variation = 1.18%) than conventional methods. Flow measurements with the technique was not significantly affected by ROI size variation, measuring position, and flow obliquity not exceeding 30°. The volume flow rates in the ICAs of a volunteer were measured with high reproducibility (coefficient of variation = 2.89% on the right, 1.48% on the left), and the flow measurement was not significantly affected by ROI size variation. The ODFE technique can minimize the effect of the non-linear background due to Maxwell terms. The technique allows use of ROIs of approximate size including the flow signal and provides accurate and objective phase-contrast flow measurements.

Key words: phase-contrast flow measurement, background of velocity-encoded image, Maxwell term phase errors, concomitant gradient fields, magnetic resonance imaging

Introduction
The phase contrast technique allows the non-invasive evaluation of flow rates and the cine phase-contrast mode has been applied to studies of various vascular systems and the cerebrospinal fluid system.\(^{1,4,6-10,12,14,15,19,21,22}\) This technique is based on the theory that the phase shifts of moving spins along a magnetic field gradient are proportional to the flow velocity. The phase shifts of moving spins are extracted by eliminating the signals of stationary spins using a pair of bipolar gradients. The flow-induced regional phase shifts obtained are converted to velocities and assigned to pixels of the so-called velocity-encoded image (velocity image), in which flow direction and velocity are represented by positive or negative pixel values.\(^{18,23}\) The volume flow rate (VFR) is determined by multiplying the mean velocity derived from a region of interest (ROI) including the flow signal in the velocity image and the ROI area. Although this method has a sound theoretical basis, some problems remain in the practical procedure. The gravest problem is that pixels without moving spins have non-zero values, and this undesirable background leads to inaccurate and poorly reproducible flow measurements. Previous studies have attributed the non-zero values of pixels without moving spins to various factors, including susceptibility effects, microcirculation effects, residual eddy current effects, and mechanical instability.\(^{5,6,14,18-20}\) However, little attention has been paid to the concomitant magnetic fields (Maxwell terms), which are considered to be a significant causative factor of the background phase shifts (Maxwell term phase errors), and few methods have been proposed to minimize the influence on phase-contrast flow measurements.

The present study developed a new method termed the “opposite directional flow-encoding
(ODFE) technique,” which is effective for minimizing the influence of the Maxwell term phase errors on phase-contrast flow measurements. In this technique, the VFR is calculated from the difference of two ROI values derived from two velocity images obtained by reversing the flow-encoding direction, which is achieved by swapping the polarity of the electric current activating the gradient. This technique frees the operators from the irritating procedure of setting ROIs exactly on the flow signals, and provides accurate, objective, and reproducible results. The validity of the ODFE technique was proved by flow-phantom experiments, and the efficacy of the technique in clinical settings verified by measuring the volume blood flow rates of the internal carotid arteries (ICAs) in four healthy volunteers.

**Theory**

A linear gradient cannot be activated without undesirable concomitant magnetic fields that give rise to a spatially dependent phase shift. The resultant phase shift causes a non-linear background which stains the velocity images, and thus affects the phase-contrast flow measurements. The concomitant magnetic field \( B_E \) is expressed as

\[
B_E(x,y,z,t) = \frac{1}{2B_0} \left\{ (G_x^2 + G_y^2)z^2 + G_z^2 \frac{x^2 + y^2}{4} - G_x(zG_y + G_yz) \right\}
\]

where \( B_0 \) is the main magnetic field oriented in the z direction, \( x, y, \) and \( z \) denote the magnet-based coordinates, and \( G_x, G_y, \) and \( G_z \) are the applied gradients. Assuming that the scan plane is limited to that prescribed by \( z = 0 \), the concomitant magnetic field is expressed as

\[
B_E(x,y,z,t) = G_z \frac{x^2 + y^2}{8B_0}
\]

In this case, the spatially dependent erroneous phase accumulation \( \phi E \) is given as the product of the integral of \( B_E \) and the gyromagnetic ratio \( \gamma \).

\[
\phi E = \gamma \int \frac{G_z(x^2 + y^2)}{8B_0} dt
\]

This equation indicates that the spatially dependent phase error accumulation is distributed symmetrically in the form of a quadratic function in the plane, and is never affected by the polarity of \( G_z \). Consequently, if the scan plane is limited to that prescribed by \( z = 0 \) and the flow-encoding gradient is applied along the z-axis, the two velocity images obtained by reversing the polarity of \( G_z \) have the same phase errors even when asymmetric flow-encoding lobes are used in each phase difference operation (Fig. 1). The ODFE technique is based on the fact that the two velocity images obtained have the same background induced by the Maxwell terms but opposite flow-encoding directions.

Conventional phase-contrast flow measurements obtain the VFR by multiplying the mean velocity derived from a ROI including the flow signal in a velocity image by the ROI area. The ODFE technique estimates the flow from two velocity images obtained at \( z = 0 \) by swapping the polarity of \( G_z \). Half of the difference between the two values derived from two ROIs placed on spatially corresponding positions of the two velocity images, which we call the “corrected mean velocity” (CMV), is substituted for the conventional mean velocity, and is multiplied by the ROI area to obtain the VFR (Fig. 2). In the cine phase-contrast mode, the CMV is calculated for each time frame and the mean CMV through a cardiac cycle is multiplied by the ROI area to obtain the volume flow in unit time (Fig. 3). With this technique, the background caused by the Maxwell term phase errors in the ROI is theoretically eliminated and more accurate flow measurements can be achieved.

**Materials and Methods**

I. **Machine and pulse sequence**

Phantom experiments and the studies on healthy human volunteers were carried out using a whole body magnetic resonance system operating at 1.5 T (Signa Advantage; General Electrics Medical Systems, Milwaukee, Wis., U.S.A.). A head coil was used for all imaging.

The phantom experiments used a radiofrequency-spoiled gradient-recalled acquisition in the steady-state (spoiled GRASS) sequence with flow-encoding gradients, and velocity images were obtained using repetition time of 50 msec, echo time (TE) of 10 msec, flip angle of 20°, slice thickness of 5 mm, field of view (FOV) of 22 cm, and matrices of 256 × 192. The study of human volunteers used a spoiled GRASS sequence with flow-encoding gradients and retrospective peripheral gating to obtain 16 velocity images in a cardiac cycle. The same imaging parameters were used as in the phantom experiments except for TE of 9.7 msec. All velocity images were acquired in the axial plane prescribed by \( z = 0 \) and the flow-encoding gradient was always applied along the z-axis.

The ODFE technique requires two velocity images with opposite flow-encoding directions, so two con-
Fig. 1 Schematic drawing of the Maxwell term phase errors included in two velocity images obtained by reversing the polarity of $G_z$, assuming that the scan plane is limited to that prescribed by $z=0$ and the flow-encoding gradient is applied along the $z$-axis. The shaded rectangles represent the phase error caused by the corresponding lobe of $G_z$. Since the phase error is a function of the self-squared term of $G_z$, the residual phase errors in the first image ($\phi E_1$) equal those of the second image ($\phi E_2$), even if the first lobe of the bipolar gradient (B) is played concurrently with the rephasing lobe (R) of the slice-selecting gradient (S).

Fig. 2 Principle of the opposite directional flow-encoding technique. ROI A and ROI B are regions of interest placed on spatially corresponding positions of the two velocity images to include the flow signals. The two ROIs include the same background, so half of the difference between the two values derived from them gives the true mean velocity (corrected mean velocity) free from background noise due to Maxwell terms. The volume flow rate is obtained by multiplying the corrected mean velocity by the ROI area.

II. Phantom experiments

The ODFE technique was evaluated by various experiments using a flow phantom connected with a free-fall circulation system. The phantom consisted of a piece of bovine artery with an inside diameter of $4.5 \pm 0.2$ mm packed in coagulated gelatin. Distilled
77

Fig. 3 Flow measurements of the human internal carotid artery (ICA) with the opposite directional flow-encoding technique. With a cine phase-contrast mode, 16 pairs of velocity images including both ICAs are acquired by reversing the polarity of Gz. The corrected mean velocity is calculated from the values measured for each pair of time frames. The mean of the 16 corrected mean velocities is multiplied by the region of interest (ROI) area to determine the volume flow rate.

water at 22 ± 2°C was filled in the system and continuously circulated. The VFR through the system was maintained at around 170 ml/min, and the actual VFR was determined using a stopwatch and a digital balance for each experimental flow measurement.

The accuracy and the reproducibility of the ODFE technique were evaluated by repeating flow measurements of the phantom. The phantom was placed in the center of the magnet and adjusted so that the long axis of the artery was parallel to the z-axis. The scan plane was determined to include the vertical cross-section of the artery, and two velocity images were consecutively obtained by reversing the polarity of Gz. A velocity-encoding value (VENC) of 150 cm/sec was used in this study. Round ROIs including 134 pixels (99 mm²) were placed on the spatially corresponding positions of the two velocity images to include the flow signals, and the difference of the two ROI values was halved to obtain the CMV. The VFR of the phantom was determined by multiplying the CMV and the ROI area. These scans, ROI measurements, and the flow rate calculation were repeated 10 times by five different operators. For simple presentation, the ratio of the estimated VFR to the actual VFR (volume flow rate index: VFRI) was calculated for each measurement. The accuracy of the technique was assessed by calculating the absolute percentage errors [(estimated VFR - actual VFR) × 100/actual VFR], and the reproducibility of the technique was evaluated by determining the coefficient of variation [(standard deviation/mean) × 100].

As a comparative study, 10 velocity images with positive flow signals acquired in the previous phantom scans were evaluated by two conventional methods of phase-contrast flow measurements: Estimation of the VFR using a ROI of size and position defined visually by referring to the corresponding magnitude image (simple method: SM), and estimation of the VFR by employing the value corrected for the background derived from another ROI placed on a region without moving spins (background subtraction method: BSM). In the latter method, a round ROI including 134 pixels was used for measurements of the flow signal and the background. The ROI for the background measurement was placed on a homogeneous region adjacent to the ROI used for the flow signal measurement. The five operators who took the corresponding scans also performed the ROI settings and the flow rate calculations. The VFRI s and the absolute percentage errors (APEs) were calculated for each conventional method and were compared with those obtained by the ODFE technique using a paired t-test. The coefficient of variation (CV) was also determined to evaluate the reproducibility of each conventional method.

The effect of ROI size on flow measurements using the ODFE technique was assessed by calculating the VFRs of the phantom using three ROIs of different sizes. A pair of the velocity images obtained by the second scan (arbitrarily chosen) of the previous experiment was used for this assessment. Three ROIs including 96, 117, and 134 pixels were applied to ROI analyses and the VFR was determined for each ROI size. The ROI measurements and the flow rate calculation were repeated five times for each ROI size, and the VFRI s and APEs obtained with different size ROIs were compared using the paired t-test.

The dependence of flow measurements by the ODFE technique on the measurement position in the plane of z = 0 was investigated by scanning the

Neurol Med Chir (Tokyo) 41, February, 2001
Table 1. Results of the comparative study between the opposite directional flow-encoding (ODFE) technique and two conventional methods of flow measurements

<table>
<thead>
<tr>
<th>Method of flow measurements</th>
<th>ODFE technique</th>
<th>SM</th>
<th>BSM</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean VFRI ± SD</td>
<td>0.995 ± 0.012</td>
<td>0.863 ± 0.077***</td>
<td>0.958 ± 0.073##</td>
</tr>
<tr>
<td>Mean APE ± SD (%)</td>
<td>1.04 ± 0.68</td>
<td>13.68 ± 7.69**</td>
<td>7.09 ± 4.12*#</td>
</tr>
<tr>
<td>CV (%)</td>
<td>1.18</td>
<td>8.91</td>
<td>7.59</td>
</tr>
</tbody>
</table>

*p = 0.0009, **p = 0.0004, ***p = 0.0002 vs. ODFE technique. *p = 0.013, ##p = 0.009 vs. SM. APE: absolute percentage error, BSM: background subtraction method, CV: coefficient of variation, SM: simple method, VFRI: volume flow rate index.

III. Studies on healthy human ICAs

The efficacy of the ODFE technique was verified in the clinical setting by measuring the VFR of the ICAs in four healthy volunteers, three males and one female aged 20–41 years (mean 31.4 years).

The reproducibility of the technique was evaluated by repeating the measurement of the VFRs of both ICAs in one of four volunteers. An axial scan plane was determined to include cross-sections of both ICAs just distal to the carotid bifurcation. Sixteen pairs of velocity images spanning a cardiac cycle were obtained by two consecutive scans with opposite flow-encoding directions. A VENC value of 80 cm/sec was used in this study. A round ROI consisted of 134 pixels (99 mm²) was placed on the spatially corresponding position of each of the 16 pairs of velocity images to include the flow signals, and the CMV was calculated for each pair of images in the same time frame. The mean of the 16 CMVs through the cardiac cycle was multiplied by the ROI area to obtain the volume blood flow in unit time (Fig. 3). Particular attention was paid to the ROI selection to prevent inclusion of the signal from the jugular veins. The scans and the flow rate calculations were repeated five times, and the CV was derived for each side to assess the reproducibility of the technique.

The effect of the ROI size on blood flow measurements by the ODFE technique was investigated by measuring the VFRs of the right ICA in three volunteers using ROIs of different sizes. The scan plane and scan parameters were the same as those of previous scans. Three ROIs including 96, 117, and 134 pixels were used for the measurements, and the VFR of the right ICA of each volunteer was determined for each ROI size. ROI measurements and the flow rate calculation were repeated five times for each ROI size. Comparisons of the VFRs obtained with different size ROIs were made for each volunteer using the paired t-test.

Results

I. Phantom experiments

The ODFE technique estimated the VFR of the phantom with mean VFRI of 0.995 ± 0.012, mean APE of 1.04 ± 0.68%, and a CV of 1.18%. All data are expressed as the mean ± standard deviation. Table 1 summarizes the results of the comparative study between the ODFE technique and the two conventional methods of flow measurements. The VFRIs were 0.863 ± 0.077 for the SM and 0.958 ± 0.073 for the BSM. The mean VFRI was significantly lower for the SM than for the BSM (p = 0.009) and the ODFE technique (p = 0.0002), whereas the difference between the BSM and the ODFE technique in VFRI was not significant. The APEs for the SM and the BSM were 13.68 ± 7.69% and 7.09 ± 4.12%, respectively. The mean APE was significantly less for the ODFE technique than for the SM (p = 0.0004) and the BSM (p = 0.0009), which indicates the highest accuracy of flow measurements with the ODFE technique. The CV was the lowest for the ODFE technique, showing the highest reproducibility of flow measurements.
Table 2  Effect of region of interest (ROI) size on flow measurements by the opposite directional flow-encoding technique

<table>
<thead>
<tr>
<th>ROI size (pixels)</th>
<th>96</th>
<th>117</th>
<th>134</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean VFRI ± SD</td>
<td>0.978±0.005</td>
<td>0.980±0.006</td>
<td>0.991±0.012</td>
</tr>
<tr>
<td>Mean APE ± SD (%)</td>
<td>2.24 ±0.53</td>
<td>2.01 ±0.57</td>
<td>1.18 ±0.85</td>
</tr>
<tr>
<td>CV (%)</td>
<td>0.55</td>
<td>0.58</td>
<td>1.22</td>
</tr>
</tbody>
</table>

No significant differences were found between the three ROI sizes (p > 0.05). APE: absolute percentage error, CV: coefficient of variation, VFRI: volume flow rate index.

Table 3  Effect of the measuring position on flow measurements with the opposite directional flow-encoding technique

<table>
<thead>
<tr>
<th>Measuring position</th>
<th>Cent</th>
<th>R-75</th>
<th>L-75</th>
<th>A-75</th>
<th>P-75</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean VFRI ± SD</td>
<td>1.004±0.018</td>
<td>1.011±0.025</td>
<td>0.993±0.018</td>
<td>1.008±0.016</td>
<td>1.002±0.027</td>
</tr>
<tr>
<td>Mean APE ± SD (%)</td>
<td>1.68 ±0.52</td>
<td>2.24 ±1.38</td>
<td>1.80 ±0.43</td>
<td>1.49 ±0.95</td>
<td>2.37 ±0.94</td>
</tr>
<tr>
<td>CV (%)</td>
<td>1.79</td>
<td>2.44</td>
<td>1.80</td>
<td>1.59</td>
<td>2.65</td>
</tr>
</tbody>
</table>

No significant differences were found between the five measuring positions (p > 0.05). A-75: 75 mm anterior to the center, APE: absolute percentage error, Cent: center of field of view, CV: coefficient of variation, L-75: 75 mm left to the center, P-75: 75 mm posterior to the center, R-75: 75 mm right to the center, VFRI: volume flow rate index.

Table 4  Effect of flow obliquity on flow measurements with the opposite directional flow-encoding technique

<table>
<thead>
<tr>
<th>Flow obliquity</th>
<th>0°</th>
<th>15°</th>
<th>30°</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean VFRI ± SD</td>
<td>1.004±0.025</td>
<td>0.995±0.032</td>
<td>1.014±0.019</td>
</tr>
<tr>
<td>Mean APE ± SD (%)</td>
<td>2.15 ±1.06</td>
<td>2.62 ±1.66</td>
<td>1.58 ±1.71</td>
</tr>
<tr>
<td>CV (%)</td>
<td>2.46</td>
<td>3.20</td>
<td>1.87</td>
</tr>
</tbody>
</table>

No significant differences were found between the three conditions (p > 0.05). APE: absolute percentage error, CV: coefficient of variation, VFRI: volume flow rate index.

Table 2 shows the effect of ROI size on flow measurements by the ODFE technique. The VFR of the phantom was measured with mean VFRI of 0.978 ± 0.005, 0.980 ± 0.006, and 0.991 ± 0.012, and the mean APEs were 2.24 ± 0.53%, 2.01 ± 0.57%, and 1.18 ± 0.85%, using ROI size of 96 pixels (71 mm²), 117 pixels (86 mm²), and 134 pixels (99 mm²), respectively. The CV tended to increase as the ROI size increased, but the mean VFRI and the mean APE were not significantly affected by ROI size.

The values of VFRI and APE obtained with the ODFE technique at five different positions in the plane prescribed by z = 0 are shown in Table 3. The mean VFRI was 1.004 ± 0.018 at Cent, 1.011 ± 0.025 at R-75, 0.993 ± 0.018 at R-75, 1.008 ± 0.016 at A-75, and 1.002 ± 0.027 at P-75. There were no statistical differences in the mean VFRI between the five measuring positions. The mean APE was 1.68 ± 0.52% at Cent, 2.24 ± 1.38% at R-75, 1.80 ± 0.43% at L-75, 1.49 ± 0.95% at A-75, and 2.37 ± 0.94% at P-75. Statistical analysis showed no significant differences in the mean APE between the five measuring positions.

The effect of the angle between flow direction and flow-encoding axis (flow obliquity) on flow measurements with the ODFE technique is summarized in Table 4. The mean VFRI was 1.004 ± 0.025, 0.995 ± 0.032, and 1.014 ± 0.019 measured when the flow...
Table 5 Volume flow rates (VFRs) in the internal carotid arteries (ICAs) of a volunteer obtained by five consecutive measurements with the opposite directional flow-encoding technique

<table>
<thead>
<tr>
<th>Flow measurement</th>
<th>VER (ml/min)</th>
<th>Right ICA</th>
<th>Left ICA</th>
</tr>
</thead>
<tbody>
<tr>
<td>1st</td>
<td>228.6</td>
<td>235.0</td>
<td></td>
</tr>
<tr>
<td>2nd</td>
<td>217.1</td>
<td>235.0</td>
<td></td>
</tr>
<tr>
<td>3rd</td>
<td>213.3</td>
<td>231.4</td>
<td></td>
</tr>
<tr>
<td>4th</td>
<td>217.8</td>
<td>234.8</td>
<td></td>
</tr>
<tr>
<td>5th</td>
<td>213.1</td>
<td>227.1</td>
<td></td>
</tr>
<tr>
<td>Mean ± SD</td>
<td>218.0 ± 6.3</td>
<td>232.7 ± 3.4</td>
<td></td>
</tr>
<tr>
<td>CV (%)</td>
<td>2.89</td>
<td>1.48</td>
<td></td>
</tr>
</tbody>
</table>

CV: coefficient of variation.

Table 6 Volume flow rates (VFRs) in the right internal carotid arteries of three volunteers measured with the opposite directional flow-encoding technique using regions of interest (ROIs) of various sizes

<table>
<thead>
<tr>
<th>Volunteer</th>
<th>ROI size (pixels)</th>
<th>Mean VFR ± SD (ml/min)</th>
<th>CV (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>134</td>
<td>245.8 ± 4.9</td>
<td>2.01</td>
</tr>
<tr>
<td></td>
<td>117</td>
<td>244.9 ± 3.7</td>
<td>1.50</td>
</tr>
<tr>
<td></td>
<td>96</td>
<td>248.5 ± 5.1</td>
<td>2.07</td>
</tr>
<tr>
<td>B</td>
<td>134</td>
<td>231.7 ± 2.6</td>
<td>1.14</td>
</tr>
<tr>
<td></td>
<td>117</td>
<td>233.7 ± 4.1</td>
<td>1.76</td>
</tr>
<tr>
<td></td>
<td>96</td>
<td>232.9 ± 4.5</td>
<td>1.94</td>
</tr>
<tr>
<td>C</td>
<td>134</td>
<td>243.7 ± 5.4</td>
<td>2.21</td>
</tr>
<tr>
<td></td>
<td>117</td>
<td>236.0 ± 5.4</td>
<td>2.28</td>
</tr>
<tr>
<td></td>
<td>96</td>
<td>239.6 ± 4.3</td>
<td>1.80</td>
</tr>
</tbody>
</table>

No significant differences were found between the three ROI sizes in each volunteer (p > 0.05). CV: coefficient of variation.

Direction of the phantom was 0°, 15°, and 30° away from perpendicular to the imaging plane, respectively. The mean APE was 2.15 ± 1.06%, 2.62 ± 1.66%, and 1.58 ± 1.71% when the flow obliquity was 0°, 15°, and 30°, respectively. No statistical differences were found in mean VFRI or mean APE between the three conditions. This demonstrates that flow obliquity not exceeding 30° has no significant influence on flow measurements with the ODFE technique.

II. Studies on healthy human ICAs

The VFRs measured in the ICAs of a normal volunteer by five consecutive measurements with the ODFE technique are shown in Table 5. The mean VFR in the right and the left ICAs was 218.0 ± 6.3 and 232.7 ± 3.4 ml/min with CV of 2.89% and 1.48%, respectively.

The VFRs in the right ICAs of three volunteers measured with the ODFE technique using ROIs of different sizes are shown in Table 6. In all three volunteers, no significant differences were found between the mean VFRs derived from ROIs of different sizes.

Discussion

Phase-contrast flow measurements may provide inaccurate and poorly reproducible results because the pixels of a velocity image without moving spins have non-zero values and the pixels at the vessel boundary include both stationary spins and flowing spins. Various methods have been proposed to improve the accuracy and the reproducibility of the measurement, including objective methods of determining the flow signal boundary and methods of eliminating the influence of linear background due to eddy current effects.5,11,14,17–19) Poor reproducibility due to inconsistent ROI definition might be improved by objective detection of vessel boundaries, but the accuracy will not be improved because the ROI includes a background bias. Phase correction using linear fitting of the background is effective for minimizing the influence of eddy current effects, but cannot eliminate non-linear background caused by Maxwell terms. An easy method employed by many investigators is to correct the ROI value using the background value derived from a neighboring region known to be free from moving spins.4,6,7,9,13,22) This method may yield passable results when the measurement position is near the center of the magnet. However, the accuracy will considerably deteriorate when the vessel of interest is in the periphery of the magnet, because the background is not linear but quadratic. In contrast, the ODFE technique effectively eliminates the influence of the non-linear background caused by the Maxwell term phase errors, improving the accuracy and the reproducibility of phase-contrast flow measurements.

In the present study, the ODFE technique could estimate the VFR of the phantom with high accuracy (mean APE = 1.04%) and high reproducibility (CV = 1.18%). The comparative study showed the mean APE was significantly lower using the ODFE technique than using the SM (p = 0.0004) and the BSM (p = 0.0009). The CV was lowest for the ODFE technique suggesting the highest reproducibility of
flow measurements. These results indicate that the ODFE technique can minimize the influence of the background on phase-contrast flow measurements most effectively and can extract the signal from moving spins most accurately.

The great advantage of the ODFE technique is that ROIs of approximate size can be used including the flow signal in velocity images. Our phantom study revealed that ROI size variation had no significant effect on the flow measurements. Since the ODFE technique is effective in eliminating the spatially dependent non-linear background in the plane prescribed by \( z = 0 \), the volume flow rate ought to be estimated accurately wherever the ROI is placed in this plane. This was verified by the phantom study, which showed no significant differences between the VFRIs obtained at five different positions in the plane.

Pixels including moving spins are not excluded from the ROI, but the flow measurement is not greatly affected by vessel obliquity because the decrease in mean velocity is canceled by the increase in vessel area. As mentioned before, the ODFE technique allows a generous ROI setting. This advantage of the technique prevents the mistake of excluding pixels with moving spins even if partial volume effects obscure the vessel boundary. Our phantom study demonstrated that a flow obliquity not exceeding 30° has no significant influence on the flow rate measurement.

Our study revealed that the advantages of the ODFE technique verified in phantom experiments were also valid in flow measurements of human arteries with pulsatile flow. Repeated flow measurements in the ICAs of a volunteer showed high reproducibility of the technique (\( CV = 2.89\% \) on the right, \( 1.48\% \) on the left), though the accuracy was not evaluated because a more accurate method of flow rate measurement is not available. Flow measurements of the right ICAs in three volunteers demonstrated that the ODFE technique was not significantly affected by ROI size variation.

Drawbacks of the ODFE technique are the requirement for two scans for one flow measurement and the scan plane is limited to the axial plane prescribed by \( z = 0 \). However, this is the only technique to minimize the influence of the Maxwell term phase errors on phase-contrast flow measurements without remodeling gradient waveforms and the reconstruction program. A new method of correcting the concomitant phase shifts by passing the information about gradient waveforms to the reconstruction program has been proposed. I expect that the ODFE method will also work well in phase-contrast flow measurements.

Acknowledgment

The author wish to thank Professor Kaoru Kurisu and Dr. Masayuki Sumida of the Department of Neurosurgery at Hiroshima University School of Medicine (Hiroshima), Dr. Takayuki Suzuki and Dr. Koichi Fujikawa of the Department of Medical Imaging at Hiroshima General Hospital, and Dr. Tohru Uozumi, Professor Emeritus of Hiroshima University, for their technical support and great help in preparing the manuscript.

This paper was presented at the 21st Annual Meeting of the Japan Society for CNS Computed Imaging in Sapporo in February 1998.

References

2) Bernstein MA, Zhou X, King KF, Ganin A, Pelc NJ, Glover GH: Shading artifacts in phase contrast angiography induced by Maxwell terms: Analysis and correction. Presented at the International Society for Magnetic Resonance in Medicine; April 12, 1997; Vancouver, British Columbia, Canada

Neurol Med Chir (Tokyo) 41, February, 2001


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**Commentary**

The author has developed a new method termed the opposite directional flow-encoding (ODFE) technique. This technique is effective for minimizing the influence of the Maxwell term phase errors on phase-contrast flow measurements, by correcting the non-linear background of velocity images induced by concomitant magnetic fields. The author emphasizes the higher accuracy and reproducibility of this method for phase-contrast flow measurement and the advantages of this technique were proved by various flow-phantom experiments. The efficacy of this technique in clinical settings was also verified by measuring the volume blood flow rates of the internal carotid arteries in 4 volunteers. With this technique, flow measurement is not significantly affected by ROI size variation, measuring position, and flow obliquity not exceeding 30 degrees. Therefore, this technique allows use of ROIs of approximate size. Finally, this idea is useful for neurosurgeons who have many opportunities to measure the blood flow and also very helpful to the radiology technician by eliminating the irritating procedure of exact ROI setting.

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The author tried to correct the artifacts that affect flow velocity measurement with magnetic resonance imaging. He focused on the artifacts caused by concomitant magnetic fields, which are called Maxwell term phase errors, and developed a new method termed the opposite directional flow-encoding technique. The validity of this method was assessed with phantom experiments and trials in human volunteers. The data obtained are reasonable and accurate. The method would be useful for researchers who are interested in physiological carotid flow recruitment during task or pathological changes in patients. The author is congratulated for developing this method.

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