Rapid Measurement of Pulse Wave Velocity via Multisite Flow Displacement

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A MR method is presented for measuring pulse wave velocity (PWV) and its application to assessing stiffness in the human thoracic aorta. This one-dimensional (1D) flow displacement method applies a single RF comb excitation to the vessel, followed by an oscillating frequency encoding gradient, each oscillation providing a 1D projection of the vessel, enabling one to track fluid motion. The currently implemented sequence excites nine slices within a 20-cm length of vessel and has a temporal resolution of 2.03 msec and a total acquisition time of 140 msec. Offline-reconstructed position-versus-time plots show curvilinear flow displacement trajectories corresponding to fluid motion at each of the excitation positions. The PWV can be reliably calculated by curve-fitting these trajectories to a model. In vitro studies using compliant tubes demonstrate no significant difference between results obtained using this method and those directly obtained using pressure transducers. Compared to another MR method previously developed in our laboratory, the proposed method displays improved temporal resolution and enhanced ability to extract PWV from vessels exhibiting low peak flow velocity. Preliminary data suggest that this method is feasible for in vivo application and may provide a more accurate estimation of aortic wave velocity among subjects exhibiting low peak flow velocity, such as the elderly or those with impaired cardiac function.

Currently, cardiovascular disease is the leading cause of death in the United States (1). Aortic stiffness has been shown to be a powerful independent indicator of cardiovascular risk (2–5). The consequences of aortic stiffening may include elevated systolic blood pressure and pulse pressure (6), increased left ventricular afterload (7), and impaired coronary perfusion (8,9). Thus, an accurate, non-invasive assessment of aortic stiffness potentially has significant clinical value (10).

One of the most reliable indices of vessel stiffness is the pulse wave velocity (PWV), the rate of propagation of the pressure or flow wave. PWV is predicted by the Moens–Korteweg equation,

\[
PWV = \frac{Eh}{2\rho r^3}.
\]

where \(E\) and \(h\) are Young’s modulus and thickness of the vessel wall, respectively, \(r\) is the vessel radius at end diastole, and \(\rho\) is blood density. In practice, the PWV is usually obtained by measuring the time for the pulse wave (pressure or flow wave) to travel a certain distance along a blood vessel. In humans, aortic PWV may range approximately from 3 to 15 m/sec with a high value indicating a stiffer vessel. A number of methods, e.g., carotid–femoral PWV using applanation tonometry (11) or ultrasound (12), have been employed to access aortic stiffness. However, such measurements are not localized to the aorta, and in addition, the vessel length can only be approximated, thereby compromising accuracy. Certain advantages of MR (deep tissue penetrability and precise spatial positioning) allow a localized assessment of aortic PWV with potentially increased accuracy.

Proposed MR methods for PWV measurement include phase velocity encoding (13), Fourier velocity encoding (14,15), and real-time acquisition and evaluation (16). Alternatively, by measuring vessel cross-sectional area and flow at different cardiac phases, aortic PWV can also be assessed (17). The major difficulties with these methods include lengthy acquisition times, insufficient temporal resolution, and susceptibility to cardiac gating irregularities. More recently, a MR-tagging technique has been described (18), which allows a PWV measurement to be completed in a single cardiac cycle. However, the spatial resolution is low, and the success of this method is dependent on the aorta being straight over a significant length, which is often not the case among middle-age and older subjects.

The one-dimensional (1D) velocity method previously developed in our laboratory (19–21) is a similarly rapid PWV measurement. It employs repeated radiofrequency (RF) tagging and time-of-flight tracking of aortic blood motion at just two (upstream and downstream) sites, separated by 84 mm. Validation experiments have shown that this method may provide reliable PWV measurements both in vitro and in vivo. From extensive studies in human subjects, however, we have found that this method may suffer impaired precision when peak systolic blood velocity within the aorta is relatively low. This occurs due to an inherent trade-off between sequence temporal resolution and fluid velocity resolution and is exacerbated by residual signal from static tissue. For this method, there is a fixed relationship between the RF pulse repetition time (TR) and the echo time (TE), whereby TE is roughly equal to TR plus the acquisition time. The trade-off occurs because a shorter TR is desired for higher temporal resolution, whereas a longer TE aids in distinguishing low-ve-
locity blood at early systole from zero velocity background signal. As a result, in subjects whose systolic blood velocity is low (e.g., less than 50 cm/sec), relatively few data points are obtained during the systolic upstroke, thus increasing the uncertainty of the PWV estimation.

This article presents an alternative strategy, which we call a “1D displacement method,” for acquiring 1D fluid displacement trajectories for improved PWV measurement. In contrast to the repetitive two-site RF tagging strategy described above, this new sequence uses a single multifrequency (9-tooth comb) RF pulse to excite aortic blood in late diastole, followed by an oscillating frequency encoding gradient to track fluid flow throughout one systolic period. The result is improved temporal and fluid velocity resolution, thereby more clearly revealing the initial part or “foot” of the systolic flow waveform, and increasing measurement accuracy among subjects whose peak aortic flow velocity may be relatively low. Moreover, the proposed method does not require that the thoracic aorta be straight and allows correction of PWV based on the curvature of the vessel.

METHODS

All phantom and in vivo experiments were conducted using a clinical 1.5-T MR system (Vision, Siemens Medical Solutions, Erlangen, Germany) equipped with 25 mT/m gradients. A spine array surface coil was used for signal reception.

Multifrequency RF Pulse

To excite multiple thin parallel slices simultaneously, a specially designed pulse superficially similar to a BURST RF pulse was implemented (22,23). In our proposed sequence, we employ a pulse that excites nine equally spaced planar slices within a field of view of 20 cm. This comb pulse was constructed as follows: two cosine functions ($s_1$ and $s_2$) of different frequencies ($f_1$ and $f_2$) but identical amplitudes and phases were selected. Identical durations of the two functions $s_1$ and $s_2$ were alternately concatenated together. A Hamming window was used to shape the final RF pulse envelope (Fig. 1a). The corresponding frequency domain spectrum is shown in Fig. 1b, where the frequency range of −28.3 to 28.3 kHz corresponds to a vessel segment of 20 cm. The peak amplitudes are 45 dB above the baseline noise, on average. Each excitation peak is thin (corresponding to a spatial width of 1.5 mm at 40 dB and about 3 mm at 0 dB), in order to maintain sharply defined fluid displacement profiles. The interval (in Hz) between two adjacent peaks is determined by the frequency of occurrence of the alternating $s_1/s_2$ functions in the time domain. The relative amplitude of each excitation peak is shaped by $f_1$ and $f_2$. In this study, the following design parameters were chosen: duration of $s_1$ and $s_2$ 75 μsec each, $f_1$ 5 kHz, $f_2$ 17 kHz, pulse duration 3.2 msec, number of support points 640.

MR Sequence

A timing diagram of the MR sequence is shown in Fig. 2. The sequence is initiated by the electrocardiogram R-wave trigger. The first segment is an initial variable delay, followed by a presaturation segment, and finally the rapid PWV measurement segment. The variable delay (35 to 95 msec) accommodates subject-to-subject variation in the arrival of the flow wave at the first upstream measurement site. Next, one or more spatial presaturation slabs is applic...
plied to partially suppress excess signal from static tissue adjacent to the thoracic aorta. During the measurement segment, slice selection and frequency encoding are deployed parallel to the vessel orientation (nominally the Z direction). No phase encoding is employed, and all signals are 1D projections along the vessel long axis. Immediately following RF excitation, the read gradient is oscillated, and an echo is acquired after each gradient reversal (every 2.03 msec). The total measurement time is 140 msec. Currently implemented sequence parameters are as follows: receiver bandwidth 125 kHz, analog-to-digital converter sampling time 1024 μsec, number of acquired echoes 64, slice selection gradient 6.7 mT/m, read gradient 14.7 mT/m, corresponding to a 20-cm field of view.

Data Acquisition and Analysis

Since software to analyze the collected data is not commercially available, a customized program dubbed wave velocity analysis (WVA) was implemented on a personal computer using MatLab (The MathWorks, Inc., Natwick, MA). WVA allows flow images to be displayed in color, which enhances visualization of flow traces. The program also provides several useful tools to facilitate wave velocity analysis. For example, flow signals can be automatically or manually tracked, and, using mathematical model fitting, the wave velocity can be automatically calculated.

The flow image was constructed from the FFT of the time domain raw data sets. Zero-padding produced a flow image matrix of 64 by 4097 temporal and spatial pixels, respectively. Each temporal pixel represents 2.03 msec, and each spatial pixel represents 0.05 mm. To accommodate the selected FOV of 20 cm, three elements of the spine array coil were activated during data acquisition. Each element generated one data set and each data set consisted of 64 acquired echoes. The final flow image was assembled during postprocessing by replacing the left and rightmost 5-cm margins from the middle element with data from the adjacent elements. This step increased the signal-to-noise level at the periphery of the flow image. Flow signals can be observed in these images as bright traces leaving their original static positions along the flow direction. The PWV can be extracted from analysis of these flow traces.

Two methods of data analysis have been developed and used for this study. The first is a curve fitting method, used for both in vitro and in vivo studies. It adopts a mathematical model, which is similar to that introduced by Macgowan et al. (18), but with additional compensation for aortic curvature (Fig. 3). The mathematical model can be expressed by Eqs. [2–4]. Equation [2] is based on the assumption that each flow trace can be partitioned into two segments: a constant velocity section and a constant acceleration section, which correspond to residual diastolic flow and systolic accelerating flow, respectively. Equation [3] is based on the second assumption that PWV is constant within the measured field of view. Equation [4] is introduced to correct for aortic curvature. The PWV calculation uses nonlinear model fitting of all the discernable flow signals with a least-square algorithm.

\[
\begin{align*}
\text{PWV}(t_n - t_l) &= h_n + \frac{v_n \cdot t_l}{\cos(\theta_n)} - \frac{v_n \cdot t_l}{\cos(\theta_n)} \\
&= h_n + \frac{d_n}{\cos(\theta_n)} - h_{n-1} \quad 2 \leq n \leq N \quad [3]
\end{align*}
\]

Here, \( N \) is the total number of visible traces; \( n \) represents the current flow trace number, with 1 indicating the most proximal visible trace; \( s_n(t) \) is the current displacement of trace \( n \) at time \( t \); \( v_n \), \( a_n \), and \( t_n \), respectively, represent the initial velocity, systolic constant acceleration, and the time of initial systolic acceleration; \( d_n \) (see Fig. 3) is the distance from the first visible trace to trace \( n \); \( h_n \) is the actual vessel length between the initial positions of trace 1 and trace \( n \) and \( \theta_n \) is the angle between the read gradient direction and the local aortic orientation within adjacent trajectories \( n \) and \( n + 1 \), and this angle can be measured from the sagittal scout image. The duration of constant acceleration is constrained to be less than 36 msec.

The model fitting process includes the following steps. First, find \( h_n \) from Eq. [4]. Next, use \( h_n \) in Eq. [3] to find \( t_n \) with respect to the unknown PWV and then insert \( t_n \) into Eq. [2] and fit \( s_n(t) \) for each given \( t \). A total of \( 2N + 2 \)
parameters can be obtained from the curve fitting process. They are \( \text{PWV}, t, v_n (n \text{ from } 1 \text{ to } N) \), and \( a_n (n \text{ from } 1 \text{ to } N) \).

The second method to calculate PWV involves converting displacement trajectories to velocity curves. This method was used only in vitro to compare with the results obtained from the curve fitting method. From a given flow displacement trajectory, the instantaneous blood velocity at each time point can also be determined. This was implemented using a rolling temporal window of width 8.12 msec and iteratively calculating the net displacement over each interval. The instantaneous velocity is this displacement divided by 8.12 msec. From the instantaneous velocities, velocity curves can be constructed. These velocity curves not only are useful for determining peak blood velocity, duration of constant acceleration, and other quantities, but also can be utilized to extract PWV.

One selects a segment having a relatively constant velocity increase and fits these points via least-square linear fitting. The intersection of this line with the time axis can be regarded as the flow foot (13). This is done for each trajectory. Finally, from a plot of position vs. time, least-square linear fitting is again applied through all measured flow feet. The slope of this line represents the pulse wave velocity. In this study, since this method was applied only in vitro, vessel curvature correction for PWV calculation was not considered.

In Vitro Validation

To validate this MR PWV measurement method, two latex tubes of differing compliance were employed. The first was relatively stiff, with 19 mm inner diameter (ID) and 3.2 mm wall thickness. The second tube had a 25.4 mm ID and 0.3 mm wall thickness and was more compliant. The tubes were sequentially placed in a fluid circuit consisting of an 8-liter tank containing tap water and a pulsatile pump (Model No. D141, Liquid Metronics, Acton, MA) in a closed-loop configuration. In several experiments, a 4-cm-deep water bath was placed under the latex tubes to simulate residual static tissues in vivo. For the in vitro MR measurements, the tubes were oriented parallel to the magnet Z axis. PWV results using the proposed MR method were compared with those measured using two pressure transducers. For the stiffer tube, two short 23-G hypodermic needles, separated by 0.8 m, were directly inserted through the tube wall. For the compliant tube, the needles were inserted immediately distal to the two rigid connectors at either end of the soft tube. Since the pulse propagation time within the short connectors can be neglected, the effective separation for calculating PWV was taken as the distance between the proximal edges of the two connectors. The needles were attached to pressure transducers (Transpac II, Abbott Laboratories, Chicago, IL), whose amplified outputs were recorded by a two-channel digital oscilloscope (Model No. 54622A, Agilent Technologies, Palo Alto, CA) at a sampling rate of 4000 Hz. Each digitized pressure waveform was numerically analyzed to determine the foot, i.e., the time point at which the pressure just began to increase. The foot was defined as the intersection between the time axis and a straight line tangential to the point of maximal derivative on the pressure curve. The effective separation divided by the time delay between the feet of the pressure curves yielded the PWV. Fifteen measurements were conducted for each tube to assess reproducibility.

In Vivo Comparison with Previous Method

Five healthy human subjects, 22 to 72 years of age, underwent aortic PWV measurements using the proposed MR method and the 1D velocity sequence previously described (19). The study was approved by the VCU institutional review board, and all participants gave informed consent. Subjects were positioned supine on a spine array receiver coil and centered at the level of the xiphisternum. Spatial presaturation was applied coronally on either side of, and parallel to, the descending thoracic aorta. For each subject and each method, PWV was defined as the average of seven measurements.

RESULTS

In Vitro Experiments

Figure 4 shows sample images constructed from the MR data collected from the stiff tube. The horizontal axis represents spatial position and the vertical axis is time (increasing from bottom to top). Eight flow displacement traces are clearly visible in both Fig 4a and b. In Fig. 4b, the
observed vertical straight lines reflect the signals from the static water bath. The total distance from the rightmost trace (upstream) to the leftmost (downstream) is 19 cm and the thickness of the flow trace at the initial time is about 3 mm. The temporal resolution is 2.03 msec. From the direction of the displacements, it is clear that the fluid motion is toward the right. By automatically tracing the displacement trajectories (with some manual correction required at flow onset), numerical data were acquired for curve fitting with the model to calculate PWVs. Likewise, using calculated velocity curves from the flow displacements, PWVs can also be extracted in addition to peak flow velocities for each trajectory. Figure 5 illustrates an example of this latter method from the data of Fig. 4b. The symbols represent the calculated instantaneous velocity for one representative trajectory. The boldface line represents the slope for constant velocity increase (constant acceleration) for this particular trajectory. The flow feet are marked by circles. The peak flow velocity for each measurement was taken as the average of the eight peak flow velocities corresponding to each trajectory.

Table 1 shows the results of pulse wave velocity measurements, using both pressure data and MR data, for the two latex tubes. Using one-way ANOVA, the measured PWV by MR using either data analysis method was not significantly different from the corresponding dual pressure measurement, regardless of the presence of static signal (stiff tube: $f = 0.73$, $P < 0.05$; compliant tube: $f = 1.28$, $P < 0.05$). A very small SD was achieved for the 1D displacement measurement in the absence of static signal. The coefficient of variation was about 5% for both tested tubes. When static water was introduced, this coefficient of variation increased to about 9%. In humans, aortic PWV typically ranges from 4 to 10 m/sec and it would be unlikely to observe a PWV above 16 m/sec. Thus, the selected latex tube models are representative of both a typical human aorta and the upper limit of in vivo PWV.

**In Vivo Experiments**

Figure 6 is a sample image from one of the human subjects (male, age 46). Eight flow displacement traces (rightward intensity curves) are visible. In addition, residual background static signals (vertical lines) and flow signals from other vessels (weaker signals to right or left) can also be seen. Due to the interfering static signals, a fully automatic flow signal detection scheme was not used. Instead, by first manually tracing the flow trajectories using a trackball, the PWV could be calculated (see Subject 3 in Table 2).

![FIG. 5. PWV calculation based on calculated velocity curves. The velocity points (star symbols) were calculated by differentiating one representative displacement curve of Fig. 4b. (For clarity, other velocity points are omitted.) The curved line is the result of a fourth-order polynomial fit of the velocity points before the peak. The sloping straight lines represent linear fits of the constant acceleration segments of all eight displacement curves. Intersections of these lines with the time axis denote the flow “feet” (marked by circles) for each trajectory. The calculated PWV for this trial was 15.7 m/sec.](image)

![FIG. 6. Aortic flow displacement image from a 46-year-old healthy male subject. In addition to the aortic blood traces, residual static signals can also be observed as lines parallel to the time axis. The superimposed oblique straight line intersects the flow feet, which were identified by nonlinear model fitting to the digitized flow signals. The PWV is 5.9 m/sec; the mean residual for each trajectory is 0.6 mm.](image)

<table>
<thead>
<tr>
<th>Latex tube</th>
<th>Dual pressure measurement PWV (m/sec)</th>
<th>MR displacement measurement</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Method 1*</td>
<td>Method 2**</td>
</tr>
<tr>
<td></td>
<td>PWV (m/sec)</td>
<td>PFV (m/sec)</td>
</tr>
<tr>
<td></td>
<td>No static</td>
<td>With static</td>
</tr>
<tr>
<td>1 (stiff)</td>
<td>15.83 ± 0.15</td>
<td>15.7 ± 0.6</td>
</tr>
<tr>
<td>2 (compliant)</td>
<td>5.52 ± 0.05</td>
<td>5.7 ± 0.3</td>
</tr>
</tbody>
</table>

**Note.** PWV, pulse wave velocity; PFV, peak flow velocity.

*Direct model fitting of fluid displacement trajectories.

**Using velocity curves calculated from displacement trajectories.
The comparative wave velocity results from both our conventional velocity sequence and the new displacement method in four of the subjects are given in Table 2 (Subjects 1 to 4). Peak blood velocity (PBV) for each subject was obtained from the 1D velocity method. For the velocity method, fourth-order polynomial curve-fitting of the velocity points achieved high fitting quality ($R^2 > 0.97$) and the coefficient of variation of PWV was 3–10%. For the displacement method, at least five flow trajectories were curve-fit to a model for each PWV determination. The mean residual (difference between observed displacements and the mathematical model for each trajectory) for the four subjects was 0.6 mm. The PWV coefficient of variation was 9–13%, which is slightly higher than that of the velocity method. However, there was no significant difference in PWV for the two methods.

The fifth subject in Table 2 was a 72-year-old woman. The PBV measured by the velocity method was extremely low, with a mean value of 0.45 m/sec. Since fewer velocity points were resolved during the early systolic upstroke for both downstream and upstream, the fitting quality was poor ($R^2 < 0.90$). As a consequence, PWV could not be reliably estimated using the 1D velocity method. Conversely, using the displacement method, PWV was successfully estimated to be 9.8 ± 1.1 m/sec with mean residual equal to 0.7 mm (seven trials).

Subjects 1 through 4 exhibited only a small degree or extent of aortic curvature over the 20-cm field of view. Their PWV results were identical with or without curvature correction. However, the fifth subject’s aorta showed significant curvature over the entire measurement segment. Without correction, PWV was underestimated by 0.8 m/sec.

### DISCUSSION

We have described a 1D projection MR technique for measuring flow wave velocity by analyzing multisite displacements of fluid. From an in vitro validation study, we found no significant difference between results using this method (with either curve fitting or velocity curve analysis methods) and a conventional pressure PWV measurement. Preliminary trials suggest that in vivo measurements of aortic wave velocity are feasible using this MR sequence along with the curve fitting data analysis method. Compared to the 1D velocity method previously developed by our group and the MR-tagging method (18), the proposed method may improve the ability to measure aortic PWV when peak blood velocity is low or the thoracic aorta is curved, as is typical among the elderly.

Comparing the two data analysis methods presented in this paper, the velocity curve method has the advantages of less computation and provides a visual means to evaluate the PWV, since the flow velocity can be observed. In vitro, it yields results consistent with other methods as shown in Table 1. At present, one problem with this method is that it is less robust for analyzing in vivo data, partially due to the inaccuracy of tracing flow trajectories quickly using a trackball. Nevertheless, this method still may provide a useful estimate of PWV in vivo.

In vitro, there is no significant difference between using eight trajectories or just two adequately spaced trajectories. Two trajectories separated by 9.5 cm yielded PWVs of 15.9 ± 0.3 and 5.7 ± 0.3 m/sec in the stiff and soft tubes, respectively, using the velocity curves data analysis method, compared with PWVs of 16.0 ± 0.3 and 5.8 ± 0.3 m/sec using eight trajectories. However, in vivo, accuracy may be enhanced by using more than two traces. One drawback of applying only two excitations is that the measurement may fail if one or both flow traces is of low quality. Therefore, the “optimal” number of excitations probably depends on overall signal quality; two are sufficient for in vitro work, but more excitations are helpful in vivo. Our preliminary in vivo study shows that a nine-excitation pulse yields very good results over a wide range of subject ages. However, since any pattern of excitations within the FOV could be designed, further comparisons between different numbers of excitations are warranted in vivo.

The preliminary in vivo data suggest that the proposed 1D displacement method and our previous 1D velocity method are complementary. In general, the 1D velocity method performs well with young and middle-age subjects, whose peak aortic blood velocity tends to be higher (above approximately 0.7 m/sec). In addition, data analysis is generally easier and much faster with the velocity method. However, since residual static background signal renders blood indiscernible at velocities below about 0.2 m/sec, for older subjects with peak blood velocities significantly less than 0.7 m/sec, the systolic flow upstroke is partially obscured. This results in too few points within the acceleration phase of the flow waveform and therefore causes PWV estimation difficulty. However, in such circumstances, the displacement method is a viable alternative because it benefits from the slow diastolic flow, which causes the blood signal to initially migrate from its static

### Table 2

**Comparison of PWV Measurements by 1D Velocity Method and Displacement Method (Average of Seven Trials)**

<table>
<thead>
<tr>
<th>No.</th>
<th>Age</th>
<th>Sex</th>
<th>PBV (m/sec)</th>
<th>PWV (m/sec)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>Velocity method</td>
<td>Displacement method</td>
</tr>
<tr>
<td>1</td>
<td>22</td>
<td>F</td>
<td>1.18 ± 0.03</td>
<td>4.3 ± 0.1</td>
</tr>
<tr>
<td>2</td>
<td>34</td>
<td>F</td>
<td>0.98 ± 0.03</td>
<td>5.5 ± 0.5</td>
</tr>
<tr>
<td>3</td>
<td>46</td>
<td>M</td>
<td>0.93 ± 0.04</td>
<td>5.4 ± 0.5</td>
</tr>
<tr>
<td>4</td>
<td>62</td>
<td>M</td>
<td>0.79 ± 0.02</td>
<td>6.6 ± 0.6</td>
</tr>
<tr>
<td>5</td>
<td>72</td>
<td>F</td>
<td>0.45 ± 0.02</td>
<td>N/A</td>
</tr>
</tbody>
</table>

Note. PWV, pulse wave velocity; PBV, peak blood velocity.
position. Second, long before the flow reaches its peak velocity, the accumulated displacement between each echo during early systole helps separate the flow signal from the background signal. Last, the higher temporal resolution (2.03 msec vs. 5.5 msec) enables the displacement method to achieve a higher data rate during the crucial acceleration phase of systole and thereby permit more accurate PWV estimation. Moreover, another advantage of the displacement method is that breath holding is not required, since a diastolic subtractive scan is not acquired.

In comparison with the MR-tagging method (18), the proposed method has several advantages. First, there is no requirement that the aorta be straight. In most subjects, the thoracic aorta is somewhat curved and tends to become more tortuous with age. However, since the introduced method excites multiple paraxial planes, aortic blood will always be excited. Displacement errors due to curvature of the aorta can be software corrected, based on the scout images. Second, the flow traces are very thin and therefore more accurately curve-fitted. Third, the proposed MR sequence is relatively simple and does not require rotational slice selection gradients. Last, the RF excitation period for this method is 3.2 msec, in contrast with more than 15 msec for the MR-tagging method. In the presence of residual diastolic flow, a longer excitation time may compromise tagging precision.

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REFERENCES