Rapid Aortic Wave Velocity Measurement with MR Imaging

The utility of a one-dimensional magnetic resonance (MR) sequence to rapidly and accurately measure wave velocity in vivo was evaluated. Studies were conducted in the thoracic aortas of 20 healthy subjects of varying age, and the MR method was validated in a compliant tube model. Aortic wave velocity ranged from 3.8 to 9.7 m/sec and demonstrated a positive correlation with subjects’ age. Peak blood velocity ranged from 47 to 125 cm/sec and exhibited a strong negative correlation with subjects’ age.

The repeated process of pumping blood by the heart produces pressure and flow waves. These waves travel throughout the circulatory system, undergoing transformation in shape and amplitude as they travel from central to peripheral arteries. While blood can be considered incompressible, vessel walls have elastic properties that result in finite pulse wave velocities (1). Wave velocity is the rate of propagation of the flow or pressure wave in an artery. The terms “flow wave velocity” and “pressure wave velocity” refer to physiologic variables that are identical in magnitude and differ only with respect to the parameter measured, that is, the flow or pressure wave. Wave velocity is an important indicator of arterial compliance, with stiffer vessels exhibiting higher wave velocities. Decreased arterial compliance is associated with aging, hypertension, and a variety of vascular diseases. A reliable means to measure aortic wave velocity would therefore be of substantial clinical benefit.

The most common method to determine wave velocity is obtaining pressure or flow curves at two locations along an artery and dividing the known distance between measurement sites by the transit time between the foot of each waveform (1). This can be accomplished invasively by piercing an artery with needles attached to pressure transducers or by threading catheter-mounted transducers into the vessel lumen (1,2). Noninvasive alternatives include applanation tonometry (3,4), ultrasonographic (US) techniques (5,6), and, more recently, magnetic resonance (MR) imaging. Among the latter are methods involving phase velocity encoding (7,8); pencil excitation with interleaved Fourier velocity encoding (9,10); use of a real-time acquisition and evaluation, or RACE, sequence (11); and comb excitation with Fourier velocity encoding (12). The main disadvantage of applanation tonometry is that it is restricted to superficial vessels, whereas the reliability of US data is affected by imprecision in probe placement. Most MR methods described to date have long acquisition times (up to several minutes) and are therefore susceptible to cardiac gating irregularities. Such difficulties are avoidable by using a faster sequence, which executes within one cardiac cycle (13).

The purpose of this study was to implement a rapid one-dimensional MR method for determining wave velocity in the thoracic aorta and to verify its accuracy in a compliant tube model.

Materials and Methods

In Vitro Studies

The MR wave velocity measurement technique was validated by using a simple compliant vessel model. This model was constructed from a 4-m length of latex tubing (inner diameter, 19 mm [3/4 inch]; wall thickness, 3.2 mm [1/8 inch]), an 8-L tank containing tap water, and a pulsatile pump (D141; Liquid Metronics, Acton, Mass) in a closed-loop configuration. Wave velocity data were obtained in the latex tube by using both the proposed MR sequence and dual-pressure measurements. The latter were performed by inserting two short 23-gauge needles...
(separated by 1 m) into the tubing. The needles were attached to pressure transducers (Transpac II; Abbott Laboratories, North Chicago, Ill) whose amplified outputs were recorded with a personal computer with a multichannel analog-to-digital converter (ACJR-12-8; Kent, Litchfield, Conn) at a sampling rate of 1,000 Hz.

Each digitized pressure waveform was numerically analyzed to determine the foot, that is, the time point at which the pressure just began to increase. This point was chosen as the intercept between the time axis and the straight line through the point of maximal derivative on the curve. The slope of this line was set to the numeric value of the maximum of the derivative of the pressure curve. The distance between the two transducers (1 m) divided by the time delay between the foot of each pressure curve yielded the wave velocity. Fifteen measurements were conducted to assess reproducibility.

In vitro MR studies were conducted with a clinical 1.5-T system (Vision; Siemens, Erlangen, Germany) equipped with 25-mT/m gradients. A spine-array radiofrequency (RF) coil with one element active was used in the experiments. The compliant tube was oriented parallel to the z axis of the magnet. A trigger signal generated by the pulsatile pump was used to initiate MR data acquisition. The general strategy of the rapid MR wave velocity technique is to reproduce the fluid flow velocity waveform at two locations along the compliant vessel (Fig 1). Because of the finite rate of propagation, the waveform arrives at the downstream location slightly later compared with the upstream location. The known distance between the two measurement sites divided by this latency yields the wave velocity. A timing diagram of the MR sequence is shown in Figure 2.

Flow-induced displacement of each RF-excited disk of fluid during the TE interval (6.5 msec) is encoded in the gradient echo as a frequency shift in direct proportion to the distance traveled. Hence, the instantaneous velocity is related to the frequency shift. Repeating the process N times (as depicted by the loop in Fig 2) allows the velocity waveform to be reconstructed with N data points, with a time resolution of 5.5 msec (the TR interval). By exciting the fluid at two locations (upstream and downstream), two velocity waveforms are simultaneously extracted from which the wave velocity can be evaluated.

In our implemented sequence, two thin sections (each 3 mm) separated by 84 mm were simultaneously excited 22 times (total acquisition time, 122 msec).
**Figure 4.** Representative spectra in a 24-year-old man (top) and a 75-year-old man (bottom) acquired near the peak of systolic flow, after zero filling and Fourier transformation. Abscissa (frequency bandwidth, 62.5 kHz) corresponds to a field of view of 100 mm; spectral resolution is 16 Hz per point or 0.4 cm/sec. Signals from both static tissues and the flow are present in each spectrum. Blood flow direction is shown by the open arrow. Note the larger separation between flow and static peak, which corresponds to higher peak blood velocity, in the younger subject.

by using a composite dual-excitation Gaussian cosine pulse. This RF pulse was generated by using the method described in reference 14. The basis for dual-frequency pulse generation is the modulation theorem (15). According to this theorem, multiplication of an arbitrary function, for which a Fourier transform exists, with a cosine function in the time domain will yield a summation of two Fourier transform replicas of this arbitrary function, for which a Fourier transform theorem (15) applies. The total duration of the saturation procedure was 35 msec. Each RF pulse was of the following form: 

\[ f(t) = C \times \text{sinc}(0.25\gamma_H \times G \times (D - d) \times t) \times \cos(0.25\gamma_H \times G \times (D + d) \times t), \]

where \( C \) is a flip angle scalar, \( \gamma_H \) is the gyromagnetic ratio for hydrogen, \( G \) is the gradient strength, \( D \) is the outer-region diameter, \( d \) is the region-of-interest diameter, and \( t \) is time. Gradients were applied in the \( x \) and \( y \) axes simultaneously (sine and cosine functions, respectively). This produced a gradient field of constant amplitude rotating about the \( z \) axis.

In our implementation, parameters were chosen to produce presaturation of a hollow cylinder (50-cm outer diameter) around an untouched cylindrical region (5-cm diameter) containing the thoracic aorta. To evaluate the degree of suppression in the outer region, this presaturation module was inserted into a standard two-dimensional gradient-echo imaging sequence and tested with a spherical water phantom, as well as in vivo.

Static signal contribution was further reduced by subtracting two data acquisitions—one containing static and flow signal, the other containing only static signal (acquired during diastole). This

**Figure 5.** Flow waveform in the latex tube (MR study). Upstream flow (○) and downstream flow (●) are shown. Polynomial fit of the fourth order with a CI of 95% was applied to the data points. R values equal 0.999 for both curves. Temporal separation between the upstream and downstream locations was evaluated at waveform midpoints. In this example, this separation is 5.10 msec, which corresponds to a corrected wave velocity of 16.0 m/sec; peak fluid velocity is 101 cm/sec.

**Static Signal Suppression**

Although the sequence in Figure 2 is sufficient for measuring wave velocity in an isolated compliant tube, its application in vivo for determining aortic wave velocity would be hindered by the surrounding static tissue. Since all tissue (not just aortic blood) in the excited sections yields MR signal, the resulting spectra will contain both signals of interest (moving blood) and signals of extraneous static tissue components. The latter contribution is, in fact, much stronger than signal from the aortic blood due to the greater mass of surrounding tissues, and, hence, the greater number of contributing protons.

Two strategies were used to reduce static signal. The first involved the suppression of signal from tissue surrounding the aorta with RF presaturation of a hollow cylindrical volume by using a method modified from those of references 16 and 17. (Standard MR regional saturation is suboptimal because of the limited width of each saturation band and the necessity to apply multiple bands manually.) Our customized saturation procedure was applied prior to the flow measurement part of the sequence (Fig 3). Sixteen saturation RF pulses were applied. The total duration of the saturation procedure was 35 msec. Each RF pulse was of the following form: 

\[ f(t) = C \times \text{sinc}(0.25\gamma_H \times G \times (D - d) \times t) \times \cos(0.25\gamma_H \times G \times (D + d) \times t), \]

where \( C \) is a flip angle scalar, \( \gamma_H \) is the gyromagnetic ratio for hydrogen, \( G \) is the gradient strength, \( D \) is the outer-region diameter, \( d \) is the region-of-interest diameter, and \( t \) is time. Gradients were applied in the \( x \) and \( y \) axes simultaneously (sine and cosine functions, respectively). This produced a gradient field of constant amplitude rotating about the \( z \) axis.
subtraction was accomplished by shifting the phases of the excitation RF pulses by 180° in the second acquisition and co-adding the raw data.

**In Vivo Studies**

This study was approved by the institutional review board; informed consent was obtained from all subjects.

Twenty healthy volunteers participated in the in vivo study (14 men, six women; age range, 22–75 years; mean age, 42 years); three (15%) of 20 subjects took part in multiday trials. The same hardware configuration was used as in the flow model studies. Subjects were positioned supine in the magnet bore by centering the laser localizers on the xiphisternum. During all data collection, subjects were asked to hold their breath on inspiration. After the acquisition of transverse and sagittal scout images, the wave velocity sequence was executed. The purpose of the scout images was to enable positioning of the wave velocity excitations perpendicular to slightly oblique aortas (typically less than 10° deviation from the z axis of the magnet), as well as to verify the correct position of thoracic aorta in the (untouched) presaturation region of interest.

Cardiac gating was performed on the QRS complex to initiate data acquisition. A variable delay (10–50 msec) between the presaturation and measurement portions of the sequence accommodated subject-to-subject variation in the arrival of the flow wave at the upstream measurement site (Fig 3).

**Data Analysis**

MR data were collected and analyzed by using spectroscopic software (Numaris 3; Siemens). Each acquired echo was stored separately for analysis. Zero filling of each of the 22 raw echoes was performed to increase the spectral resolution to 16 Hz per point in the frequency domain (corresponding to 0.4 cm/sec velocity resolution.) A one-dimensional Fourier transform was performed with each zero-filled line of raw data to obtain spectral peaks whose frequencies corresponded to the positions (along the z axis of the magnet) of the excited spins at the TE. Figure 4 shows in vivo examples of two such spectra, where causal aortic blood flow produces peaks (labeled "flow") shifted leftward from those of the RF excitation positions (labeled "static").

The frequency shift of each flow peak from its excitation position is directly proportional to the distance traveled by the excited fluid during the TE and, hence, to the fluid velocity (13). $f = f/(V_{g} \times G_{z})$ and $V_{g}$ = $Dp/TE$, where $Dp$ is fluid displacement (in meters), $f$ is spectral displacement of the particular peak relative to the excitation position (in hertz), $\gamma_{h}$ is the gyromagnetic ratio for hydrogen (in hertz per millitesla), $G_{z}$ is the gradient strength (in milliteslas per meter), $V_{g}$ is the instantaneous fluid velocity (in meters per second), and $TE$ is 6.5 msec.

Flow curves were reconstructed by plotting velocity values ($V_{g}$) versus the time following the electrocardiographic trigger, when the raw echoes were acquired. Polynomial curve fitting of the fourth order with a CI of 95% was applied to the flow points. The midpoint between zero and the peak velocity values, taken separately on each upstream and downstream fitted curve, was chosen for

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* Data are the mean plus or minus the SD.
the time-delay estimation (13). The length of the thoracic aorta between measurement sites for the majority of subjects was 84 mm (section separation distance), with few exceptions (three [15%] of 20 subjects) for nonstraight aorta segments. In those subjects, the actual length was measured on sagittal scout images.

Wave velocity was calculated as follows: \( WV = \frac{S}{T} - V_n \), where \( WV \) is wave velocity (in meters per second), \( S \) is the distance between measurement sites (in meters), \( T \) is the temporal separation between upstream and downstream waves (in seconds), and \( V_n \) is the instantaneous fluid velocity (in meters per second). The rationale for subtracting \( V_n \) in this equation was the fact that observed wave velocity is increased by the motion of the fluid itself (13). Therefore, wave velocity values obtained during the systolic part of the cardiac cycle should be corrected for the value of the instantaneous blood velocity. In cases of different \( V_n \) values for upstream and downstream waveforms, the mean value was used for correction.

An unpaired \( t \) test was performed with dual-pressure measurements and in vitro MR data to determine whether the difference between them was statistically significant.

I Results

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An example of a MR wave velocity measurement in the latex tube is given in Figure 5. A comparison of MR imaging and dual-pressure measurement–derived numeric results is given in the Table. The higher SD of the MR measurements can be largely attributed to the smaller distance between measurement sites. We deliberately chose a large separation in the dual-pressure measurement method (100 cm) to minimize the variability in timing measurements, allowing this method to serve as a standard. In the MR study, sections were separated by 8.4 cm to match this method with the in vivo methods, where there were physiologic constraints on the separation distance. Mean wave velocity values obtained with invasive dual-pressure measurement and the MR method (Table) were not significantly different.

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Examples of reconstructed waveforms in two subjects are given in Figure 7. The younger subject’s graph demonstrates a higher maximal frequency shift (higher peak blood velocity), a larger temporal separation between upstream and downstream waveform (lower wave velocity), and a sharper upstroke on the blood flow curves.

The results of the thoracic aorta study in the 20 subjects were the following: Wave velocity ranged from 3.8 to 9.7 m/sec, with a reproducibility of 13.8%, and peak blood velocity ranged from 47 to 125 cm/sec, with a reproducibility of 7.6%. Both wave velocity reproducibility and peak blood velocity reproducibility were calculated as follows: \[ \frac{\sum_{i=1}^{N} (\sigma_i/\mu_i)}{N} \], where \( N \) is the number of subjects and \( \sum_{i=1}^{N} (\sigma_i/\mu_i) \) is the sum of the ratios of the standard deviation \( \sigma \) to the mean \( \mu \) in each subject. On average, 12 trials were performed for each subject. These ranges were wide (maximum value in both wave velocity and peak blood velocity was about 2½ times greater than the minimum value), as was the age difference between the youngest and the oldest volunteers. Wave velocity measurements showed greater variability as compared with peak blood velocity measurements. In the group that took part in multiday trials, variability in both wave velocity and peak blood velocity was greater, as compared with those of subjects who participated in a single session.

Age dependence of wave velocity and peak blood velocity is shown in Figure 8a and 8b, respectively. Correlation coefficients (\( R \) values) are indicated on the graphs. Note the presence of one outlying point (in a 27-year-old man with higher wave velocity) in Figure 8a. By omitting this point, the \( R \) value would equal 0.84. In both wave velocity and peak blood velocity results, there was a

![Figure 7](image-url)
The aorta is the most attractive target for compliance measurements because aortic stiffening can initiate a cascade of serious physiologic consequences. A healthy, compliant aorta stores some of the energy of the systolic pulse by transiently distending. During diastole, elastic recoil helps maintain forward blood flow (1). Proper aortic compliance also attenuates the deleterious effect of reflected pressure waves by slowing both forward and retrograde wave propagation rates. Reduced compliance in the aorta leads to elevated impedance to ventricular ejection (cardiac afterload), higher central pulse pressure, and decreased net flow (18). Moreover, the increase in wave velocity with age is reportedly (19) greater in the aorta than in the peripheral arteries.

In this study, we experimentally evaluated the precision of a rapid one-dimensional MR sequence and conducted human trials to determine the feasibility of in vivo measurement of aortic wave and blood velocity. In vitro MR imaging and dual-pressure wave velocity results were found to agree closely (Table). Values for aortic wave velocity and its correlation with age were in agreement with the results of other similar studies (7,9,11,12).

Peak blood velocity demonstrated a strong negative correlation with age. Compared with other in vivo MR and US measurement techniques, the present study had similar (6–9,12) or better (5,11) reproducibility results. Temporal resolution was also comparable (11,12) or better (7–10).

The two main advantages of our MR method are simplicity of the measurement part of the sequence and the short acquisition time. All data were acquired within one cardiac cycle, which eliminated imprecision due to triggering irregularities and temporal data interleaving. The measurement portion of the sequence was designed to minimize TR (for improved data temporal resolution) while maintaining a reasonably long TE (to increase flow-induced spectral peak displacements). Unlike a previous one-dimensional wave velocity method (13), in this sequence we maintained an asymmetric gradient waveform to prevent re-focusing of echoes at higher multiples of the TE interval. The strong (15 mT/m) section-selective gradient effectively dephased any signal that would have appeared in the analog-to-digital converter interval immediately following each RF pulse; the gradient waveform had zero net amplitude only within the 6.5 msec TE interval.

Additional advantages of this MR method versus applanation tonometry and US techniques include the precise positioning of the measurement sites and the ability to access deep structures, such as the aorta, without the loss of signal. A more complete overview of various non-invasive methods of measuring aortic compliance can be found elsewhere (20).

Several limitations of this method warrant mention. Despite the use of static tissue suppression, the projective nature of the technique leaves a residual signal component from stationary tissue surrounding the aorta. This spectral peak can obscure flow below a certain velocity threshold and can make identification of the foot of the flow waveform difficult. Because it relies on an accurate vessel length measurement, this technique is unsuitable for assessing wave velocity in tortuous arteries. Furthermore, inadequate signal-to-noise ratio may preclude measurements in small vessels.

Less than 10% of all trials did not yield reliable wave velocity measurements, for several reasons. The most common problems were trigger-related (unstable QRS complex on the electrocardiogram or incorrect time delay estimation between QRS complex and waveform arrival into the field of view). Other complicating factors included moving heart artifacts caused by the upstream section crossing heart tissue and splitting of the spectral peak of the flow signal, especially with decelerating blood.

Accurate wave velocity assessment can provide information concerning distensibility or hardness, of arteries and has the potential for evaluation of hemodynamic function and diagnosis of a variety of vascular diseases. Rapid and accurate evaluation of aortic compliance in humans may prove valuable for studying the effects of age, exercise, or drug therapy on the vascular system.

References


