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In Vivo Validation of MR Velocity Imaging

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Abstract: Calculations of left ventricular stroke volume obtained by summing the areas of multiple contiguous transverse magnetic resonance (MR) slices in systole and diastole using a spin echo sequence have been compared in 10 healthy volunteers with the stroke output derived from velocity maps in the ascending aorta using a field even-echo rephasing sequence. The results gave a correlation coefficient of 0.97 ($p < 0.001$) and a standard error of estimate of 3.2 ml. Velocity maps have also been obtained in the pulmonary artery, the descending aorta, and the superior vena cava. The accuracy of this technique and the theoretical limitations of MR measurements have implications for the earlier detection of atheroma in the coronary and other arteries. **Index Terms:** Blood, flow dynamics—Arteries, diseases—Magnetic resonance imaging, techniques.

Magnetic resonance (MR) is a rapidly developing noninvasive method of imaging which, when performed at the levels of field and gradient strengths currently used, carries no known hazard (1–3). High resolution tomographic images with excellent contrast between tissues and blood can be obtained using cardiac gating and spin echo (SE) sequences. The images provide a detailed display of cardiac anatomy (4,5) and pathology (6–10). Patency or nonpatency of coronary artery bypass grafts may be directly inferred (11,12). Parameters of cardiac function such as chamber volume, stroke volume, and ejection fraction (13,14), wall thickness (15,16), and wall motion (17) can also be derived by analysis of the images.

Velocity mapping (18,19) using a field even-echo rephasing (FEER) sequence (20,21) provides a quantitative two-dimensional method of measuring blood velocities in the vascular system. Ascending aortic instantaneous flow for a given part of the cycle may be calculated from the velocity in each pixel in the ascending aorta as seen in transverse section; the sum of these figures at 50 ms intervals through the whole of one cycle gives a volume that should equal the left ventricular stroke volume.

We have previously shown that the sum of right and left ventricular cavity areas in multiple contiguous MR 1 cm slices taken in systole and diastole provides an accurate method of measuring stroke volume. Comparison of the figures for the right and left ventricle showed an error of only 2% in estimating the volume of a chamber (14).

Magnetic resonance thus provides two conceptually different methods for the noninvasive measurement of cardiac output. The present study reports comparison of these two methods and validation of the second.

MATERIALS AND METHODS

A Picker International Vista 2055 MR machine, operating at 0.5 T, was used for both anatomical and flow studies. Ten healthy volunteers with no previous history of cardiac disease were studied after they had given informed consent. Estimates of stroke volume were double blind, i.e., the result from one method was unknown to the person using the other.

Left Ventricular Chamber Volume

The range of transverse levels required to encompass the whole of the left ventricular cavity was established from an image in the coronal plane.

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Multiple contiguous transverse 1 cm slices were obtained in pairs, at end-systole and end-diastole, using a SE sequence with an echo time of 26 ms.

Synchronisation to the cardiac cycle was ensured by electrocardiographic (ECG) gating. End-diastole was taken as early in the cycle as the R-wave detection system would allow (20 ± 2 ms after the Q-wave, depending on the steepness of the R-wave) and end-systole at the end of the T-wave. Acquisition of a pair of such slices using two repetitions of 128 sequences took between 3 and 4 min, depending on heart rate. Eight pairs of slices usually sufficed to cover the left ventricle so that the total time required to acquire enough data was ~30 min. Image reconstruction was by two-dimensional Fourier transformation with magnitude display.

Aortic Flow

Without moving the subject, the transverse plane at the level of the right pulmonary artery was defined. Using a FEER sequence (21), up to 16 frames (depending on heart rate) were obtained at this level. The first frame corresponded to a point in the cycle 20 ms after the onset of the ECG R-wave, subsequent frames being at 50 ms intervals. Two FEER sequences were required, one flow insensitive and one sensitised to phase encode flow velocity in the slice select direction. The acquisition of these sequences was interleaved so that problems of image misregistration due to subject movement were not encountered. Two repetitions of 128 views were obtained so that the acquisition took roughly 10 min. Image reconstruction was by two-dimensional Fourier transformation for which the magnitude and phase were calculated for each pixel. Masking of the phase reconstruction to show zero velocity in pixels with a magnitude value of <10% of maximum ensured elimination of random phase values due to background noise. Subtraction of the flow insensitive from the flow sensitive phase maps resulted in phase shifts only due to the flow of blood in the slice select direction (along the axis of the ascending aorta).

Comparison of Bulk Flow with Left Ventricular Stroke Volume

Chamber Volume

The images for calculation of chamber volume were copied to an off-line computer system (Sperry-Univac V77-600) for analysis. The endocardial boundaries of each area were traced on the computer display and the computer calculated the cavity area in each slice. The areas in slices corresponding to end-diastole and end-systole were summed to obtain left ventricular end-diastolic volume and left ventricular end-systolic volume, respectively.

Bulk Flow

The ascending aorta was identified on each image and traced round on the computer display. The computer calculated both the enclosed area and the mean pixel value within the area. Since pixel values on flow maps are directly proportional to flow velocity (19), the product of these two parameters gave the instantaneous flow for the image. Stroke volume was the sum of instantaneous flows for all frames within the cycle.

RESULTS

Figure 1a shows a typical FEER magnitude image of the transverse plane through the aorta at the level of the right pulmonary artery; a velocity map for a frame in midsystole is displayed in Fig. 1b. Figure 1c shows the isolated area of the ascending aorta with a typical area defined for the measurement of mean velocity and instantaneous volume flow, and Fig. 1d displays a plot of the two-dimensional velocity profile at this point in the cardiac cycle.

Figure 2 shows the comparison of the two methods of cardiac output. The regression line of flow estimate on chamber volume estimate is $y = 0.99x + 3.2$ ($r = 0.97$, $n = 10$, $p < 0.001$, SEE = 3.3 ml).

Figure 3 shows the normal range (mean ± 2 SD) of the time course of volume flow in the ascending aortae of the 10 subjects.

Figure 4 shows measurements of the time course of ascending aortic, descending aortic, and pulmonary arterial flow in one subject.

DISCUSSION

The two methods of estimating left ventricular output are very different in concept. It is therefore reasonable to expect that they are subject to different sources of error. In chamber volume estimates there is room for subjectivity in the operator's choice of the endocardial boundary, especially in end-diastolic images when stationary blood gives a signal (22) that may obscure it. Operator experience usually limits the errors due to this problem. Additionally, the choice of inclusion or exclusion of slices near the top or bottom of the chamber in which the left ventricular wall passes almost tangentially to the imaging plane is often difficult. A coronal or a sagittal slice to determine the exact anatomical limits of the left ventricle is often helpful.

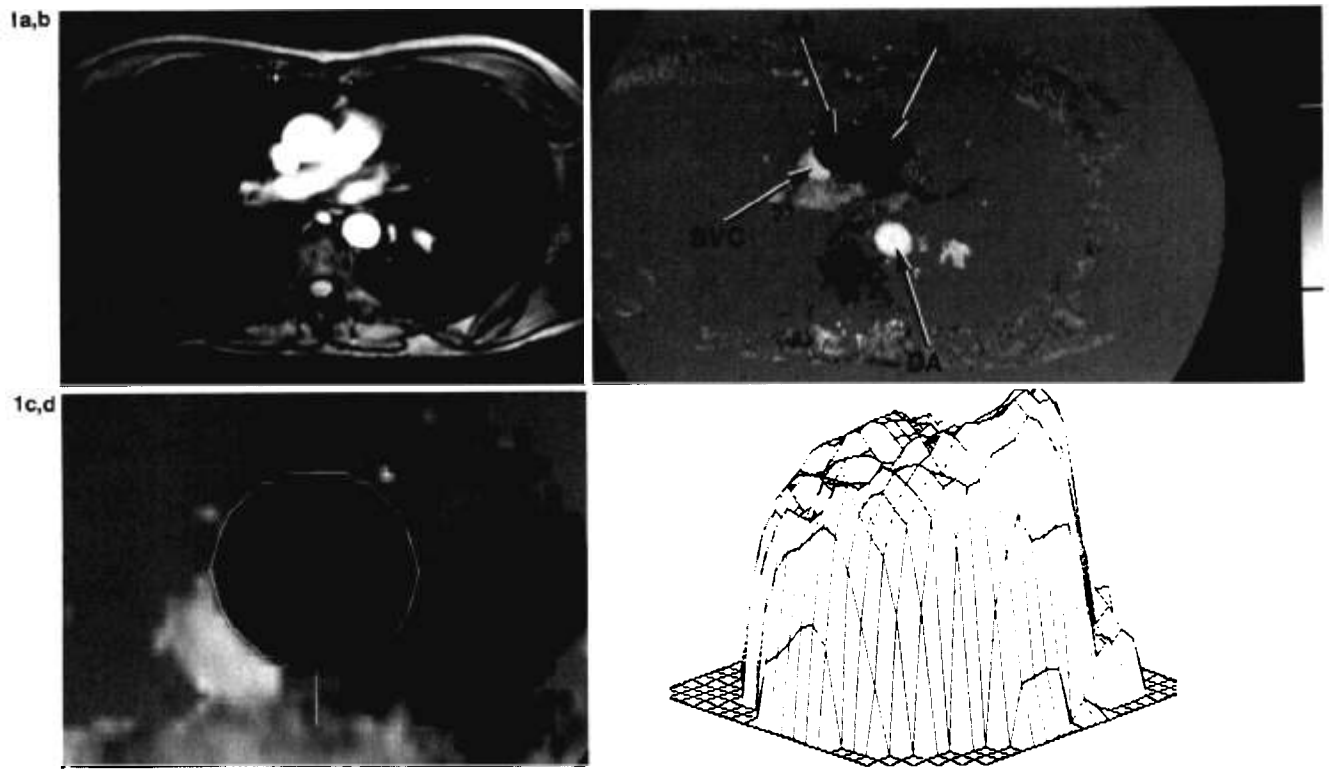

FIG. 1. a: A typical field even-echo rephasing sequence transverse section of the thorax at the level of the right pulmonary artery as used for the flow measurement in this study. **b:** Grey scale on the flow image shows zero flow as midgrey, flow towards the head as black, and flow towards the feet as white. Flow can be seen in the ascending aorta (AA), descending aorta (DA), pulmonary artery (PA), and the superior vena cava (SVC). **c:** Ascending aorta has been isolated and enlarged with an area defined for the measurement of mean velocity and instantaneous volume flow in the vessel. Each pixel has a value proportional to the velocity of flow within it. **d:** Data from the ascending aorta are displayed as a plot of the two-dimensional velocity profile at this point in the cycle.

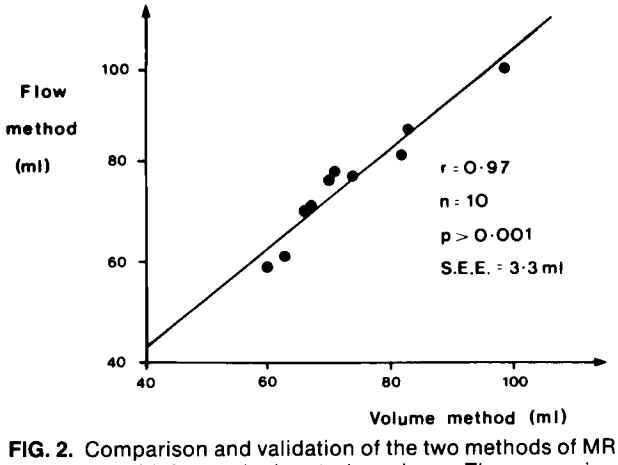

FIG. 2. Comparison and validation of the two methods of MR estimation of left ventricular stroke volume. The regression line of velocity map left ventricular output on chamber size stroke volume ($y = 0.99x + 3.2$) is very close to the line of identity. The correlation coefficient is 0.97 ($p < 0.001$) and the standard error of the estimate is 3.3 ml.

The flow method is theoretically very accurate, provided that there is sufficient signal from the flowing blood to enable phase to be calculated, that the gradient pulses that are used for phase encoding of velocity have been accurately calibrated, and that the range of flows covered has been correctly chosen. The results presented in this paper suggest that practice corresponds with theory.

In an earlier article (14) we validated stroke volume calculations by comparing right and left ventricular stroke volumes. Using these data, we have calculated the SEE for regression of right ventricular stroke volume on left ventricular stroke volume to be 6.1 ml. The SEE in this study (in which the velocity map estimate of left ventricular output was regressed on left ventricular stroke volume) was 3.3 ml. This suggests that the estimate of left ventricular output from the velocity maps is very accurate.

As well as being different in concept, the two methods are also very different in the scope of the information they provide. The chamber volume method gives anatomical information about the heart and thorax over the range of transverse levels

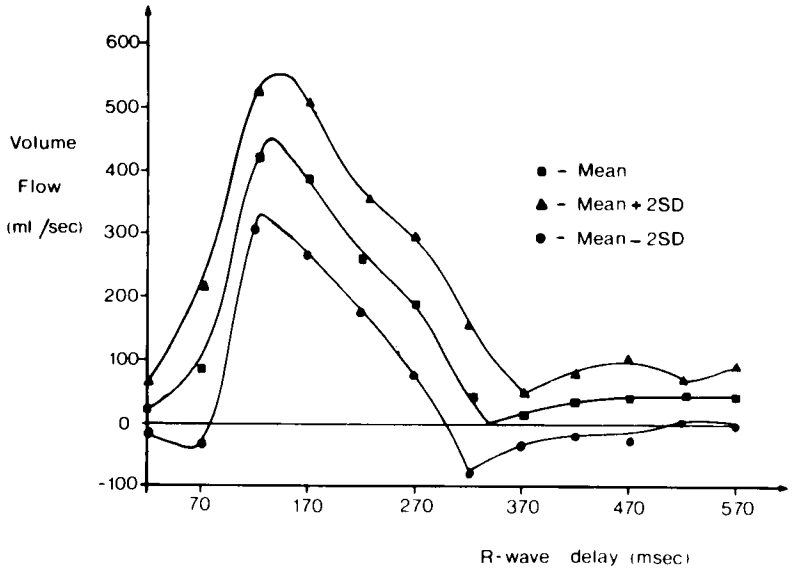

FIG. 3. The time course of ascending aortic flow in 10 healthy subjects. The upper and lower lines represent the mean + 2 SD and mean − 2 SD, respectively.

covered. Ejection fraction may be calculated, but, since the chamber volume method depends on only two points in the cardiac cycle (end-diastole and end-systole), no information about the time course of flow is available. The velocity mapping method does not provide an estimate of ejection fraction and only gives limited anatomical information (available from the magnitude reconstruction of the velocity images). Information is, however, available on the time course of aortic flow and on the details of velocity profiles. Both the time course of flow and the shape of the velocity profile are known to diverge from normal in some diseases (23). The accuracy of the two-dimensional velocity profiles is implied by their close agreement with published results obtained by hot-film anemometry in dogs (24,25) and with range gated Doppler ultrasound in animals (26) and in humans (27).

Doppler ultrasound with spectral analysis of velocities (28) can also give information on the time course of ascending aortic flow. The ordinate of the spectral analysis display gives a grey-scale histogram of the range of velocities detected within the sample volume, whereas the time course from MR velocity mapping (Fig. 3) gives instantaneous bulk flow. Ascending aortic flow is plug flow for much of systole (23), and changes in aortic diameter are relatively small making the predominant flow velocity of spectral Doppler ultrasound displays roughly proportional to instantaneous flow. Thus, the curve

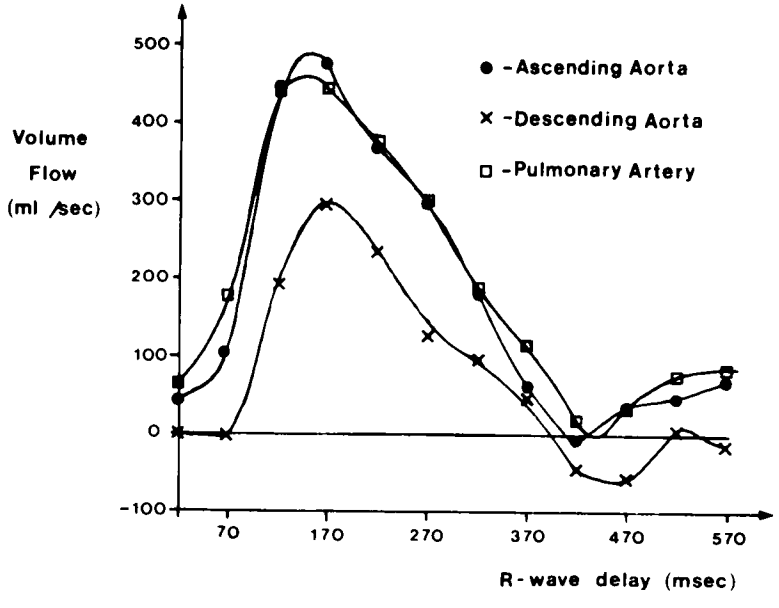

FIG. 4. Comparison of flows in the ascending aorta, pulmonary artery, and descending aorta of a healthy volunteer. See text for discussion of the relative areas under the curves.

in Fig. 3 is very similar to published ultrasound results (27). The information provided by MR velocity mapping is more complete than that from ultrasound, since two-dimensional cross-sectional maps of flow in any part of the major vessels are provided at both high spatial resolution and with several hundred steps of velocity. Ultrasound can give one-dimensional velocity profiles (26) or two-dimensional maps with only a few steps of velocity (29). With ultrasound, aliasing can sometimes make interpretation difficult at high velocities (30).

The ability of MR velocity mapping to provide velocity profiles in small vessels has already been established (19). In large vessels with very high flow velocities, signal loss due to the presence of a range of velocities within a voxel can be a problem. In the pilot work, signal loss was frequently noted in the ascending aorta in midsystole. The even echo rephasing of the FEER sequence (20,21) eliminated this problem and is thought to be an essential part of MR velocity mapping in the aorta.

The output of left and right ventricles must be equal over a few minutes. Thus the close correspondence of the areas under the curves for the ascending aorta and the pulmonary artery in Fig. 4 is to be expected. The smaller area under the curve for the descending aorta is also to be expected, since the blood for the head, the arms, and the upper thorax has left the aorta by the time this level is reached.

Further studies are in progress and MR velocity mapping is already proving to be of value in the study of congenital cardiac abnormalities in the patency of coronary artery bypass grafts (12) and in the study of vascular disease in general. The potential of MR velocity mapping to obtain velocities in a three-dimensional volume of blood in the ascending aorta is being investigated and could be of importance since flow patterns are known to change considerably with level in the aorta (25). A physical modification of the pulse sequence gives acceleration maps rather than velocity maps (31) and preliminary acceleration images have been obtained in the human aorta.

The theoretical limitations of MR in velocity mapping are such that producing velocity maps for the coronary arteries may well be within the capability of the technique. Since considerable occlusion of the lumen is needed for the appearance of symptomatic coronary artery disease, the possibility of early diagnosis at a time when pathology is not advanced enough for the risk of sudden death to be high (32) may well be possible. This might be achieved by studying abnormalities of transverse or longitudinal velocity profiles in the coronary arteries and/or acceleration. The public health implications of MR velocity mapping could therefore be enormous.

CONCLUSION

Magnetic resonance velocity mapping gives accurate estimates of total flow and velocity profiles in large vessels. The technique also provides a variety of other clinically useful information such as details of the time course of flow and velocity profiles, as well as limited anatomical information from magnitude display. The large amount of information and the short acquisition time makes it the method of choice for the noninvasive measurement of cardiac output. Its potential use in the coronary arteries could contribute greatly toward earlier diagnosis of occlusive vascular disease.

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