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Flow Measurement by Magnetic Resonance: A Unique Asset Worth Optimising

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ABSTRACT

Users and manufacturers of cardiovascular magnetic resonance (CMR) systems have, potentially, an unrivalled asset. Phase contrast mapping of velocities through planes transecting the great arteries should provide the most accurate measurements available of cardiac output, shunt flow, aortic or pulmonary regurgitation and, indirectly, of mitral regurgitation. But the reality is that phase contrast velocity mapping remains under-used, and may have become discredited in the eyes of some CMR users and referring clinicians. Even when appropriate methods of acquisition have been used, there can be inaccuracies of flow measurement on some CMR systems caused by background phase errors due to eddy currents or uncorrected concomitant gradients. Measurements of regurgitant or shunt flow can be seriously affected by these errors which should be minimised or corrected by appropriate hardware and software design. If they have not been, inaccuracies can be detected and corrected by repeating identical velocity acquisitions on a static phantom, and subtracting the corresponding apparent phantom velocities from those of the clinical acquisition. For accurate measurements of aortic regurgitation or mitral inflow, motion tracking and velocity correction with respect to the cyclic displacements of the valves are needed, but few if any commercial systems provide this facility. Measurements of jet velocity pose different challenges, mainly related to the size and placement of voxels relative to a narrow jet. Awareness of the potential problems and concerted efforts towards optimisation are needed from manufacturers and users to make appropriate use of phase contrast flow measurement—a unique strength of cardiovascular magnetic resonance.

INTRODUCTION

Phase contrast CMR measurement of blood flow has been available for over 20 years (1), and there have been publications reporting its use in a range of clinical applications. Measurements by phase contrast velocity mapping of volume flow through planes transecting the great arteries should provide the most accurate measurements available of cardiac output, shunt flow (2, 3), aortic or pulmonary regurgitation (4, 5) and, in combination with left ventricular volume or mitral inflow measurements, of mitral regurgitation (5–7), all non-invasively and without contrast agent or ionising radiation (8). But phase contrast velocity mapping remains under-used, and may have become discredited in the eyes of some CMR users and referring clinicians.

On page 681 of this issue, Chernobelsky et al. (9) address one of the most awkward and potentially serious problems. Background phase offset errors can, on certain CMR systems, lead to significant inaccuracies of flow measurement, although they may not be obvious on velocity map images. The severity of the problem can vary greatly not only between manufacturers, but also between upgrades of hardware and software, and between different planes of acquisition. The cross sectional area of the vessel in which flow is measured and the velocity encoding range set (VENC) both positively affect the amount of flow error caused by background phase offsets. As the authors point out, offset errors that seem small, perhaps only 1% of the velocity encoding range set, could cause large errors, for example 25%, in the measurement of volume or shunt flow because these measurements involve the calculation of flow through an area, integrated through the whole cardiac cycle. Even greater errors could occur if the regurgitant fraction were being calculated. We are not

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surprised by the errors in shunt measurement reported by Chernobelsky as we have been faced with similar errors and are aware that the problems are not confined to any one manufacturer.

WHAT IS CAUSING THE PROBLEM?

Many publications, including our own, have claimed accuracy for phase contrast techniques, so how widespread is the problem, and what is causing it? Our experience has been that inaccuracies were inadvertently introduced with some of the major 'improvements' of CMR performance of the last decade. In order to reduce acquisition times, the performance of magnetic gradient systems was increased in terms of the strengths and rates of change of applied gradients. In combination with advances in sequence design, this strategy has been broadly successful. Breath-hold and real-time acquisitions save time and avoid respiratory artefacts. The rapid sequences needed for breath-hold phase velocity mapping also minimise flow-related artefacts such as signal loss and time-of-flight displacement of signal (10). But an adverse effect, for velocity mapping, has been the exacerbation of non-velocity-related causes of phase offset (11). These include the presence of more significant Maxwell or concomitant gradients which are relatively well-understood and should be corrected for automatically in the software calculations of the latest generation of CMR systems (12).

CONCOMITANT OR MAXWELL GRADIENTS

When an imaging gradient is applied, the magnetic field does not only alter, as intended, linearly with position in the direction of the applied gradient, but also non-linearly due to what are known as concomitant or Maxwell fields. These concomitant fields depend on the combined gradient (G) in the x,y and z directions and their most important terms are proportional to G^2/B_0 where B_0 is the main magnetic field. Historically, with typically used fields and applied gradients, the concomitant fields were relatively small and could be ignored, but the move toward increased gradients for rapid imaging results in more significant non-linear phase variations across the image. These may be too small to be noticeable in terms of image distortion, but different phase errors between the pair of images acquired for velocity encoding can lead to significant errors in the velocity map that is then calculated by subtraction. Figure 1 shows a velocity acquisition from one of our own systems (Siemens Sonata, 1.5 Tesla) in 2001, before the implementation of correction for concomitant Maxwell gradients.

EDDY CURRENTS

After correction of errors due to concomitant gradients, the main remaining causes of background phase offsets are eddy currents. These are eddying or swirling movements of charge induced in electrically conducting components, particularly those with relatively large cross sectional areas. As in complex flow, electromagnetic eddies may not be easy to predict accurately.

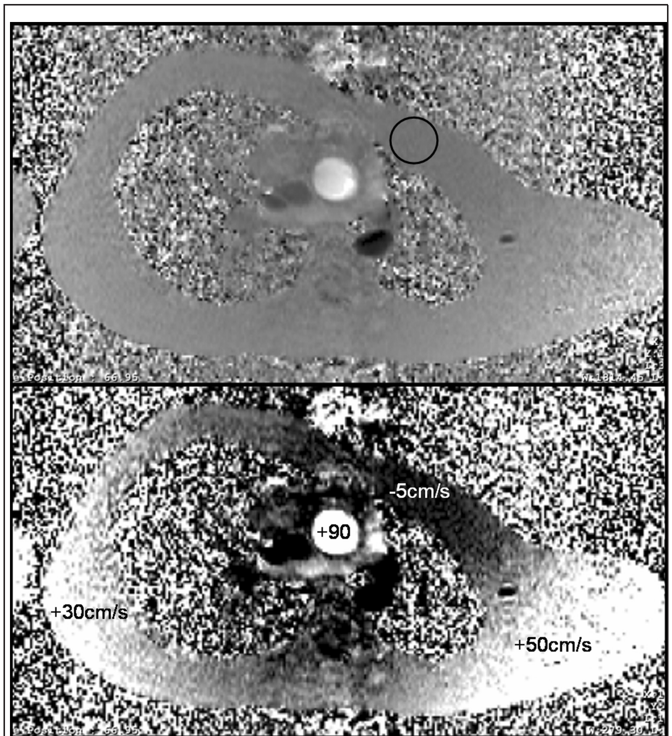

Figure 1. Severe background phase errors on a breath-hold through-plane phase contrast aortic velocity map, without any correction for concomitant gradients. The problem is not immediately obvious on the upper image, which is displayed with normal image contrast settings. Placement of a 25 mm diameter circle in signal from the static chest wall, however, (black circle) records a mean 'velocity' of −5 cm/s, and apparent 'flow' of over one litre per minute through the circle, towards the feet. The lower image shows the same velocity map with more extreme contrast, making the non-linear background phase errors apparent, with locally measured 'velocity' values indicated.

Nevertheless, principles are known by which they can be avoided or minimized, for example by active gradient shielding or by the replacement or subdivision of structural components with non-conducting materials and the use of less extreme switches of gradient. Retrospective ECG gating generally gives less errors due to eddy currents than prospective gating, which requires the cyclic interruption and re-starting of the acquisition.

CORRECTION OF FLOW MEASUREMENT USING A PHANTOM ACQUISITION

Phase offsets due to concomitant gradients and eddy currents may not be distributed linearly across the field of view, so attempted correction by the zeroing of static regions of signal, for example in the chest wall, can make matters worse rather than better in the vessel of interest. This is why the phantom correction technique used by Chernobelsky et al. (9) may be necessary and appropriate. Following a clinical study, their method is to repeat identical velocity acquisitions on a static phantom, using an ECG simulator. Applying identically located and sized areas of interest to those used for measuring flow in the great vessels, the corresponding apparent phantom 'flows' are then subtracted from those of the clinical acquisition. This approach may only take a few minutes of acquisition and post-processing time, but this adds up in a busy clinical schedule, and should not be regarded as an acceptable solution to the problem.

To identify if a particular CMR system is subject to background phase errors, it is important to start with a high level of suspicion. Phantom correction acquisitions should be run if discrepancies are noticed between measurements and expected results, for example if significant apparent shunt flow is measured in a healthy volunteer, or if a low aortic regurgitant fraction is measured in a patient with an obviously incompetent valve. This should not only correct the study in question, but give an idea of the potential significance of the problem.

BASIC PHANTOM CHECKS

Ideally, we would like to be able to recommend a standard phantom test to run on any CMR system to check its suitability for flow measurement. But it soon emerges that there are numerous variables that might modify the results. These include slice location, orientation, VENC, voxel dimensions, echo time, repetition time, gating method and even routine service engineering recalibration of eddy-current correction parameters. Nevertheless, we feel that there should be a spatial region extending to a minimum of 5 cm from the centre of the magnet, preferably to 10 cm, within which routinely used flow acquisitions are sufficiently free of background phase offsets to be used for flow measurements. This would probably mean that any apparent velocities measured within this region, through a plane located across a static phantom using an ECG simulator for gating, should not exceed, and should preferably be considerably less than 0.5% of the velocity encoding range set. For this test, mean velocities should be measured in an area of interest rather than at an individual pixel.

If a circular area of 25 mm in diameter, approximately aortic size, is placed appropriately on the phantom 'velocity' map, the apparent flow through the area should not exceed 0.2 litres/minute if the VENC had been set at 2 m/s. Such a flow error would represent about 4 or 5% of a typical adult cardiac output. We are aware that not all currently used CMR systems meet these standards, and that clinically used sequences and slice orientations vary. The most relevant slices are the oblique planes, tilted at about 45° between transaxial and coronal, and about 45° between transaxial and sagittal, which are used for aortic and pulmonary flow measurement, as illustrated in Figure 2. We ran such a test, locating the centre of 45° oblique 'aortic' and 'pulmonary' slices between 0 to 10 cm from the magnet centre towards the 'head' direction. Our 1.5 Tesla Siemens Avanto system showed only small phase errors in most cases, with a maximum tested 'flow' error of 0.2 litres/minute through a 25 mm circle. Our 1.5T Siemens Sonata system showed potentially significant errors, up to a maximum of 0.4 litres/minute using 'breath-hold' and up to 0.3 litres/minute using 'non-breath-hold' sequences.

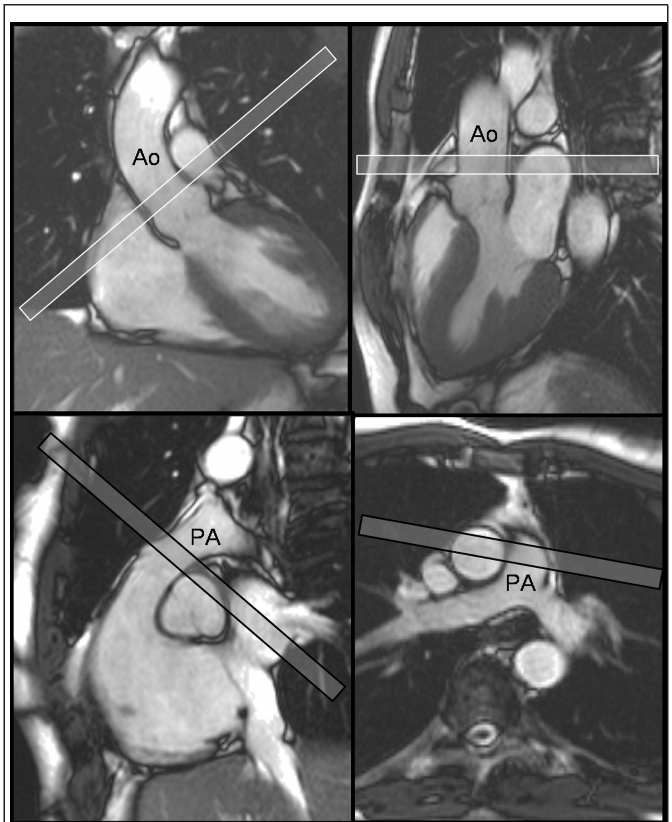

Figure 2. Suggested locations of velocity mapping planes for aortic (above) and pulmonary artery (below) flow measurements, relative to diastolic cine images. If moving slice velocity acquisition is available, it may be preferable to locate the aortic velocity mapping slice immediately proximal to the coronary origins. It is important to locate the slice for pulmonary artery flow measurement proximal the the bifurcation. Ao = aorta. PA = main pulmonary artery.

As background phase errors tend to increase with distance from the centre of the magnet, it is important to position a patient with the regions of interest in the great vessels no more than 5 cms from the centre of the magnet. The errors associated with eddy currents depend on the strengths and rates of change of the gradients used, so sequences requiring less extreme gradient switches should be less prone to background phase errors.

If significant background phase errors are found to be present, it is important that the manufacturer is informed and asked to investigate and correct, where possible, the underlying cause. Inaccuracies may be at least partially avoidable with the installation of appropriate software.

MEASUREMENTS OF AORTIC AND PULMONARY FLOW AND REGURGITATION

During a clinical study, planes of acquisition must be located appropriately with respect to the vessels in question. It has been recommended that the acquisition plane for aortic flow be located across the aortic sinuses, between the valve and the sino-tubular junction, just proximal to the origins of the coronary arteries (13). Due to the movement of the aortic root, slight variability of breath-hold positions, and without the motion correction techniques outlined below, however, this may be hard to achieve and replicate in routine clinical practice and follow-up. We therefore chose to locate the plane at or immediately above the sino-tubular junction at end diastole, as shown in Figure 2. This keeps the plane clear of any convergent, accelerating diastolic flow in the vicinity of a regurgitant orifice, but it must be understood that coronary flow, which is typically about 5% of the cardiac output (9), does not reach this plane.

On the pulmonary side, the regurgitant orifice can be wide, and complete absence of effective pulmonary valve function is relatively common after repair of tetralogy of Fallot or valvotomy for congenital pulmonary stenosis. A regurgitant fraction of about 40% is typical in patients with no effective pulmonary valve action, but it can vary considerably depending on upstream and downstream factors, not only on the incompetence of the valve itself (14).

CYCLIC MOVEMENTS OF VALVE PLANES AND CORRECTIONS FOR THEM

All four heart valves normally move with respect to the chest wall and the magnet during the cardiac cycle. Such movement may be reduced after previous surgery, for example for tetralogy of Fallot, but otherwise the displacements of valve planes can affect attempted measurements of regurgitant flow. In diastole, when an aortic regurgitant jet is flowing back into the ventricle, the root moves up in the opposite direction, typically by about 6–12 mm (Figure 3). This tends to cause underestimation of the regurgitant volume or regurgitant fraction. If the root is dilated and mobile, the underestimation of the regurgitant fraction could be as much as 10 or 15% of the forward flow volume. If unrecognised, this could give misleading evidence regarding the need for surgical intervention.

A solution to this source of inaccuracy is the implementation of motion tracking and heart motion adapted flow measurements as described by Sebastian Kozerke and colleagues in 1999 and 2000 (15, 16). This important technique has yet to be made available on most commercial CMR systems. The displacements of the aortic root or mitral annulus may be tracked by a modified tagging technique, and both the location of the velocity mapping slice *and the through-plane velocity offsets* are adjusted to take account of the annular movements through the phases of the cardiac cycle. In this way, velocities are measured relative to the valve annulus rather than relative to the magnet or the body. In their second paper, the authors showed how this procedure corrected significant underestimates of aortic regurgitant fractions in patients, and that the need for correction was even greater for accurate measurement of mitral flow. While it may not be realistic to measure mitral regurgitant flow directly due to the shape of the valve and the narrow or splayed nature of the jets that pass back through it, mitral regurgitant volume is calculable by subtracting systolic aortic outflow from diastolic mitral inflow, as long as both are measured accurately.

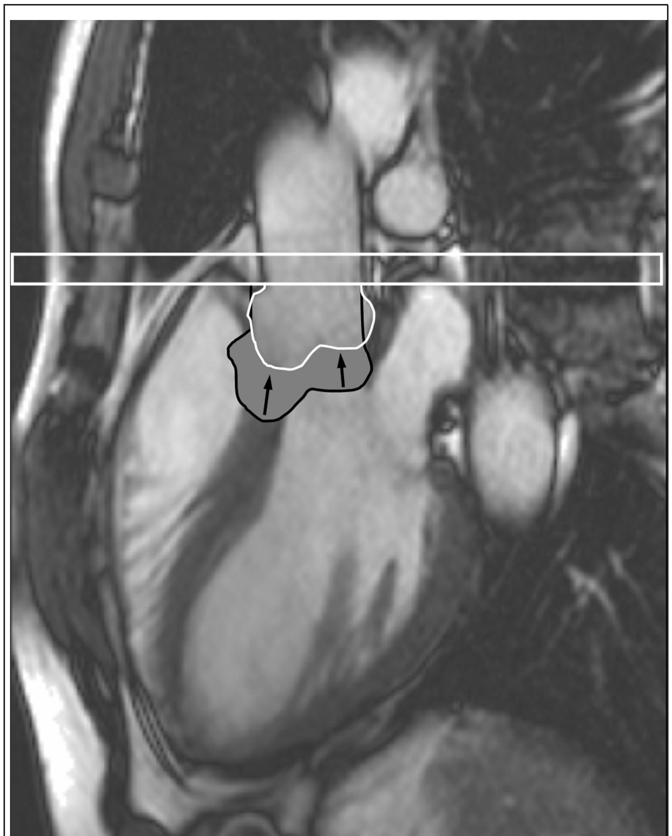

Figure 3. The displacement of the aortic root from its end systolic (curved black line) to its end diastolic (curved white line) positions in a healthy 54 year old volunteer. The white lines indicate the location of a static velocity mapping slice used for aortic flow measurement. The volume change of the root beneath the plane, indicated by the grey area, gives an equivalent volume of apparent diastolic forward 'flow' through the plane of acquisition, minus the amount of coronary flow. The volume of any regurgitant flow would be underestimated by this amount, which would be greater if the root were dilated.

JET VELOCITY MEASUREMENT BY CMR

Measurements of high velocity jet flows have their own set of challenges and limitations, different from those of measuring volume flow in the great arteries. Background phase errors are less critical as cross sectional areas are small and velocity rather than volume flow is usually measured without the need for integration through time. But a key consideration is the shape, size and orientation of voxels relative to those of the jet (Figure 4). The velocity of a jet can only be measured reliably by CMR if it has a coherent jet core (its central, high velocity, low shear region) large enough to contain entire voxels. The voxels are relatively long and thin, their length being the through-plane slice thickness. The cores of coherent jets are also long and thin, extending downstream from the orifice. Jet velocity is therefore best measured *through* a plane located carefully to transect the core of the jet, immediately downstream of the orifice. Irregular or very narrow jets, however, particularly those

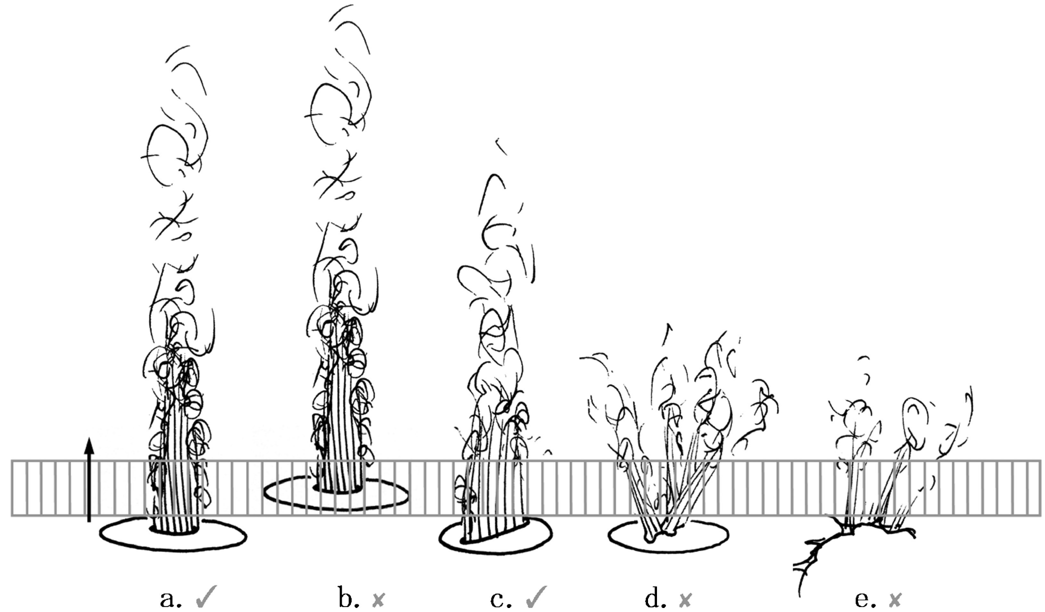

Figure 4. For jet velocity mapping, the velocity mapping slice (grey grid) should transect the coherent core of the jet immediately downstream of the orifice (a and c) so that voxels lie entirely within the jet core. The slice should not span the level of the orifice (b) as it would then include signal from blood accelerating on the upstream side. Voxels spanning the acceleration zone or the abruptly decelerating shear layer at the edge of a jet are unlikely to measure velocity accurately. If jets are fragmented, splayed, very narrow or oblique (d and e) their velocities are unlikely to be measured accurately by CMR.

of regurgitant valves or a severely stenosed aortic valve, may be fragmented and unsuitable for accurate velocity measurement by CMR, whereas Doppler ultrasound is not necessarily subject to the same limitations.

OTHER VELOCITY MAPPING APPLICATIONS

The through-plane velocity mapping technique can also be used for the sizing of atrial septal defects, ventricular septal defects and regurgitant orifices, particularly of the tricuspid and pulmonary valves. For these applications, the velocity mapping slice is located to transect the jet or stream passing through the orifice. Appropriately low VENCs are used for low velocity jets, for example 80 cm/s for an ASD stream, and 250 cm/s for sizing the defect of a severely regurgitant tricuspid valve.

With simultaneous catheter measurements of pressure, pulmonary arterial flow measurements probably allow the most accurate available calculations of pulmonary resistance, particularly in the presence of shunts (17).

A PLEA TO THE MANUFACTURERS

Manufacturers should ensure that phase velocity encoding software on existing as well as new CMR systems includes effective correction for the predictable concomitant gradients. At the design stages, hardware components of systems need to be designed to minimise eddy currents. Where existing systems are subject to eddy currents, it may be necessary to include sequences for flow measurement with lower rates of change of gradients, making it clear that the more rapid flow sequences, which may still be preferable for jet velocity measurement, are unsuitable for quantifying regurgitant or shunt flow. This may mean that free breathing acquisitions need to be used, possibly taking two minutes or more, or spatial and temporal resolution may need to be compromised, but this should be more acceptable than significant inaccuracies due to background phase offsets.

Finally, we believe that the current generation of CMR systems should allow moving slice, motion corrected velocity acquisitions as described by Kozerke et al. (15, 16). These should allow the measurement of mitral as well as aortic regurgitation with unprecedented accuracy.

To end on a positive note, flow measurements of the type discussed here represent a unique and unrivalled strength of CMR. Consistent optimisation and quality control of CMR systems for measurements of flow will be in the interests of patients, CMR users and manufacturers.

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