Session: 4D-flow: Ready for Primetime?
Title: 4D-flow: How We Acquire It?
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Highlights

- 4D-flow acquisitions require a significant amount of data
- Clinically useful protocols require the use of parallel imaging and/or compressed sensing

Target Audience

This talk is designed for physicians, clinicians, and scientists interested in understanding some of the different strategies used in reducing the scan times required for 4D-flow acquisitions.

Outcomes/Objectives

Participants should come away with an understanding of why 4D-flow acquisitions are intrinsically time consuming and a familiarity with several of the different methods that can be used to reduce scan times to clinically viable lengths.

Purpose

4D-flow refers to a time-resolved phase contrast acquisition in which velocity encoding is performed in all three spatial dimensions (1). Unlike 2D imaging, the volumetric nature makes it easier for the operator to prescribe the study without the need to be familiar with the details of the underlying anatomy. Furthermore, post-processing can provide both flow and anatomical quantification, potentially making it possible to replace multiple time- and operator-intensive 2D studies with a single exam (2).

While 4D-flow as a technique has been around since the late 90s (3), its clinical acceptance has been hampered by its long scan times. Generally, the total number N_k of needed (k_y,k_z) k-space locations are divided up into N_{seg} segments that are repeatedly sampled during one R-R interval. This means that the total number of R-R intervals needed to complete the scan is given by $T_{RR} = N_k/N_{seg}$. In addition, when prescribing a 4D-flow study the operator must also consider the underlying temporal resolution of the data. To collect flow information in all three physical dimensions, four acquisitions must be obtained at each (k_y,k_z) location (one for each flow direction and one to serve as a baseline to correct phase errors) (4). Therefore, if the repetition time of the sequence is denoted by TR, then the temporal resolution of the data is given as $T_{res} = 4\times TR\times N_{seg}$.

As an example, consider a sequence with a TR of 5 ms with which the operator would like to obtain an in-plane resolution of 128 phase encodes and 64 slices. The total number of k-space locations needing to be sampled is thus $N_k = 128\times64 = 8192$. If these are divided up into a segment length of $N_{seg} = 4$, then the total number of heart beats needed to complete the scan is $T_{RR} = 8192/4 = 2048$, and the temporal resolution of the data would be $4\times4\times5$ ms = 80 ms. While an 80 ms temporal resolution might be sufficient for many applications, one can see with a moderate heart rate of 1 beat/sec, the total acquisition time would take 2048 seconds, or just over 34 minutes. While a greater segment length could be used to reduce scan time at the expense of temporal resolution, in general this is not practical for routine clinical use.

The example above highlights another issue that has slowed the acceptance of 4D-flow imaging. Unlike 2D multi-slice acquisitions, the length of the exam makes breath holding

impossible. Therefore any acquisition scheme needs to address the potentially substantial respiration artifacts that can lead to degradation in image quality.

Methods

Respiration artifacts can be addressed in several different ways. Simple respiratory compensation can be done by selecting an appropriate set of (k_y,k_z) phase encodes based on the phase of the respiratory cycle at the instant of the R-R trigger, which is termed respiratory ordered phase encoding, or ROPE (5). In the best case the effect is a set of images that appear to have been acquired over a single long breath. While this technique tends to produce only moderate image quality improvements, no extra scan time is needed.

Navigator gating generally produces better image quality by rejecting data that are acquired outside of an acceptance window that is usually based on diaphragm position (6-8). Because this technique does not require respiratory bellows, it has the added advantage of greater patient acceptance. This approach can also be used together with ROPE to determine which set of phase encodes will be acquired when in the window so that any remaining artifacts can be minimized.

Aside from potentially increasing scan times, navigators can be problematic in that any time spent acquiring navigator data is time not spent measuring flow. One approach to solving this issue is the use of self-gating techniques to measure k-space profiles motion at regular, shorter intervals during the study (9). Data acceptance can then be based on motion estimates derived from the correlation of measured and reference profiles.

The most straight-forward approach to reducing scan times is to use a basic parallel imaging technique such as GRAPPA to reduce the number of phase encode sets that need to be acquired (10). A small number of central lines in both the k_y and k_z direction (typically on the order of 20-24) are fully sampled to be used in the generation of k-space weights. The remaining (k_y,k_z) pairs are subsampled by factors on the order of 2-3 that are determined by the coil used for the study. The k-space weights are then used to synthesize the missing data during reconstruction. GRAPPA reconstruction techniques tend to be fairly robust, but are somewhat limited in how much reduction can be used before significant artifacts begin to appear.

With the development of compressed sensing reconstructions in conjunction with pseudo-random sampling patterns such as variable density Poisson disk, the number of needed phase-encode sets can be further reduced (2,11). While in all cases the practical amount of reduction also depends on the coil geometry and number of elements, compressed sensing reconstructions will generally tolerate larger reduction factors than when using parallel imaging alone (12).

Parallel imaging techniques provide scan time reductions by reducing the number of phase encode sets that are needed for each cardiac phase. Even greater scan time reductions can be realized by exploiting the spatio-temporal correlation inherent in the dynamic data. Methods such as *k-t* BLAST and *k-t* SENSE can achieve 5 to 8-fold increases in acceleration by varying the undersampled sampling pattern from phase to phase (13,14).

Finally, non-Cartesian acquisitions such as spiral and radial can exploit their increased sampling efficiency to reduce scan times and improve resolution (15,16). Three-dimensional radial sampling is advantageous in that it can provide spherical isotropic resolution at the millimeter level in scan times on the order of 7-12 minutes (17,18). Non-Cartesian applications are prone to aliasing issues from objects larger than the in-plane FOV, but unlike in Cartesian

sampling the resulting artifacts are less structured and more benign in appearance.

Results

As was seen above, scan times for fully sampled, moderate resolution datasets can be on the order of tens of minutes, making 4D-flow impractical for regular use. Parallel imaging techniques such as GRAPPA have now been along for close to ten years and as such are well accepted in the clinic. With an appropriate coil, outer reduction factors of 2 are easily possible in both the in-plane and slice directions when collecting data. In previous work we've shown examples of how this approach can be used to reduce the acquisition time of a ($0.9 \times 1.1 \times 2$) mm^3 data set from 2412 to 552 R-R intervals (or 40 to 9.2 minutes assuming a 60 bpm heart rate) (19).

Newer approaches to data reduction using compressed sensing reconstructions such as L_1-SPIRiT can reduce these scan times even further (20). Using 32 channel body coils, our institution typically uses overall (that is, both in-plane and slice direction) compressed sensing reduction factors of 9-11 to image volumes with (1.3 x 1.6 x 2.6) mm^3 in approximately 5 minutes, and sub-millimeter in-plane resolutions of (0.8 x 0.8 x 2.6) mm^3 in approximately 9 minutes.

Discussion

In addition to data reduction strategies to reduce scan times, 4D-flow imaging will also benefit from the use of contrast agents to improve signal-to-noise ratios (SNR). In the past gadolinium-based agents have been used due to their availability and the fact that 4D-flow sequences often are performed after angiographic studies (MRAs). However, because of the ensuing delay between injection and data acquisition, gadolinium-based agents can tend to wash out and become less effective. More recently, blood-pool agents such as gadofosveset and ferumoxytol have shown promise in providing high SNR while maintaining their concentration over the course of the study.

4D-flow acquisitions can also benefit from the addition of fat saturation (21). Because the lipid signal is very bright and tends to originate from close to the chest wall, eliminating this signal can help in reducing the severity of the resulting respiration artifacts. This can also help improve vessel delineation and make segmentation easier during data processing. The additional pulses will lengthen the TR somewhat and decrease temporal resolution, but if desired this can be recovered by decreasing the segment length N_{seg}.

Conclusion

While 4D-flow as a technique has been in development for close to 20 years, its clinical acceptance has been limited by the long scan times necessary for data acquisition. The advent of both parallel imaging and compressed sensing techniques in conjunction with high-channel count coils has made it possible to obtain data sets with millimeter-level resolutions in times on the order of 5 minutes. Clinical acceptance will now depend on vendor adoption and the development of better software to visualize the data.

References

1. Markl M, Chan FP, Alley MT, Wedding KL, Draney MT, Elkins CJ, et al. Time-resolved three- dimensional phase-contrast MRI. J Magn Reson Imaging. 2003;17(4):499–506.

2. Hsiao A, Lustig M, Alley MT, Murphy M, Chan FP, Herfkens RJ, et al. Rapid pediatric cardiac assessment of flow and ventricular volume with compressed sensing parallel imaging volumetric cine phase-contrast MRI. Am J Roentgenol. 2012;198(3):W250–9.

3. Wigström L, Sjöqvist L, Wranne B. Temporally resolved 3D phase-contrast imaging. Magn Reson Med. 1996;36(5):800–3.

4. Pelc NJ, Bernstein MA, Shimakawa A. Encoding strategies for three‐direction phase‐contrast MR imaging of flow. J Magn Reson Imaging 1991;1:405-413.

5. Bailes DR, Gilderdale DJ, Bydder GM, Collins AG, Firmin DN. Respiratory ordered phase encoding (ROPE): a method for reducing respiratory motion artefacts in MR imaging. J Comput Assist Tomogr. 1985;9(4):835–8.

6. Ehman RL, Felmlee JP. Adaptive technique for high-definition MR imaging of moving structures. Radiology. 1989;173(1):255–63.

7. Firmin D, Keegan J. Navigator echoes in cardiac magnetic resonance. J Cardiovasc Magn Reson. 2001;3(3):183–93.

8. Baltes C, Kozerke S, Atkinson D, Boesiger P. Retrospective respiratory motion correction for navigated cine velocity mapping. J Cardiovasc Magn Reson. 2004;6(4):785–92.

9. Uribe S, Beerbaum P, Sørensen TS, Rasmusson A, Razavi R, Schaeffter T. Four-dimensional (4D) flow of the whole heart and great vessels using real-time respiratory self-gating. Magnetic Resonance in Medicine. 2009;62(4):984–92.

10. Bammer R, Hope TA, Aksoy M, Alley MT. Time-resolved 3D quantitative flow MRI of the major intracranial vessels: initial experience and comparative evaluation at 1.5T and 3.0T in combination with parallel imaging. Magn Reson Med. 2007;57(1):127–40.

11. Lustig M, Donoho D, Pauly JM. Sparse MRI: The application of compressed sensing for rapid MR imaging. Magn Reson Med. 2007;58(6):1182–95.

12. Vasanawala SS, Alley MT, Hargreaves BA, Barth RA, Pauly JM, Lustig M. Improved pediatric MR imaging with compressed sensing. Radiology. 2010;256(2):607–16.

13. Baltes C, Kozerke S, Hansen MS, Pruessmann KP, Tsao J, Boesiger P. Accelerating cine phase- contrast flow measurements using k-t BLAST and k-t SENSE. Magn Reson Med. 2005;54(6):1430– 8.

14. Jung B, Stalder AF, Bauer S, Markl M. On the undersampling strategies to accelerate time- resolved 3D imaging using k-t-GRAPPA. Magn Reson Med. 2011;66(4):966–75.

15. Gu T, Korosec FR, Block WF, Fain SB, Turk Q, Lum D, et al. PC VIPR: a high-speed 3D phase- contrast method for flow quantification and high-resolution angiography. Am J Neuroradiol. 2005;26(4):743–9.

16. Sigfridsson A, Petersson S, Carlhäll C-J, Ebbers T. Four-dimensional flow MRI using

spiral acquisition. Magnetic Resonance in Medicine. 2012;68(4):1065–73.

17. Frydrychowicz A, Landgraf BR, Niespodzany E, Verma RW, Roldán-Alzate A, Johnson KM, et al. Four-dimensional velocity mapping of the hepatic and splanchnic vasculature with radial sampling at 3 tesla: a feasibility study in portal hypertension. J Magn Reson Imaging. 2011;34(3):577–84.

18. Frydrychowicz A, Landgraf B, Wieben O, François CJ. Images in Cardiovascular Medicine. Scimitar syndrome: added value by isotropic flow-sensitive four-dimensional magnetic resonance imaging with PC-VIPR (phase-contrast vastly undersampled isotropic projection reconstruction). Circulation. 2010;121(23):e434–6.

19. Alley MT, Beatty PJ, Hsiao A, Vasanawala SS, Reducing the scan time of time-resolved, 3D phase contrast imaging with 2D autocalibrated parallel imaging, in *Proc., ISMRM, 18th Annual Meeting*, Stockholm, page 1343, 2010.

20. Murphy M, Alley M, Demmel J, Keutzer K, Vasanawala SS, Lustig M, Fast l_1-SPIRiT Compressed Sensing Parallel Imaging MRI: Scalable Parallel Implementation and Clinically Feasible Runtime, IEEE Trans. Med. Imaging 31, 1250 (2012).

21. Alley MT, Hsiao A, Vasanawala SS, Reduction of Respiration Artifacts in 3D Phase Contrast Imaging with Intermittent Fat Saturation, in *Proc., ISMRM, 20th Annual Meeting*, Melbourne, page 1195, 2012.