Quantification of Turbulence Intensity by Generalizing Phase-Contrast MRI

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Introduction

Blood flow in the normal vascular system seems to be remarkably free of turbulence. Turbulent flow, however, is believed to be implicated in the pathogenesis of several cardiovascular diseases. For example, studies indicate that turbulence is involved in the initiation and development of atherosclerosis [1].

Turbulent flow causes signal loss in phase-contrast magnetic resonance imaging (PC-MRI) due to the spin velocity distribution within a voxel during the influence of a magnetic field gradient. Pipe has presented a method for estimating the standard deviation (SD) of velocity within a voxel [2]. In this work, we extend that method and present a theoretical derivation of a method for direct measurement of the SD of the blood flow velocity distribution within a voxel. For validation, we quantify the turbulence intensity (TI) in post-stenotic flow and compare our findings with previously reported laser Doppler anemometry (LDA) measurements.

Theory

The theory is based on the analytical expression modeling the signal of a voxel in presence of a bipolar gradient, $S(k_v) = C\int_{-\infty}^{\infty} s(v)\cdot\exp(-i\, k_v\, v)dv$, where C is a scaling factor encompassing receiver gain, relaxation effects and spin density. s(v) is the spin velocity distribution within the voxel and $k_v = \gamma M_1$, where γ is the gyromagnetic ratio and M_1 is the first gradient moment. In an ordinary PC-MRI velocity acquisition, k_v for the two required measurements is related to the velocity encoding range (VENC) according to $\mathrm{VENC} = \pi/(k_{v1}-k_{v2})$. By assuming that s(v) has a Gaussian distribution and considering two measurements with different first gradient moment, the signals from the measurements can be related to each other according to $S(k_{v1}) = S(k_{v2})\cdot\exp(-\sigma^2(k_{v1}^2-k_{v2}^2)/2-i(k_{v1}-k_{v2})v_m)$ where v_m is the mean flow velocity. Taking the magnitude and rearranging gives $\sigma = \sqrt{(2\ln(|S(k_{v2})|/|S(k_{v1})|)/(k_{v1}^2-k_{v2}^2))}$ which states that the SD of the velocity distribution within a voxel, in one direction, can be calculated from the signal *magnitude* of two PC-MRI measurements of different first gradient moment. For best precision, one of the measurements should be acquired using zero first gradient moment and the other with a first gradient moment adapted to the expected SD. After measuring the SD, the TI is obtained by $TI = \sigma/v_m$ [-] according to its definition.

Methods

For validation, we performed an in-vitro comparison between our results on TI and previously published LDA results [3]. We used a flow phantom consisting of a 14 mm diameter Perspex® tube with a 75 % area reduction cosine shaped stenosis (fig. 1a) and a 63 % glycerol solution maintained at a temperature of 33 °C giving it a kinematic viscosity of 0.12 cm^2s^{-1}. 3D phase-contrast data sets (TR = 15 ms, α = 20°, FOV = 130x97.5x160 mm, matrix size 64x48x80) were acquired at three flow settings characterized by different Reynolds numbers (Re) (table 1) using a clinical 1.5 T scanner. The pulse sequences were optimized for measuring SD, i.e. zero first gradient moment was used in one of the measurements. In addition, a velocity measurement (VENC = 4 m/s, TE = 3.1 ms) was carried out for Re = 1000.

Table 1. Flow and imaging parameters.

Reynolds number [-]	500*	1000*	2000*
Mean flow velocity [m/s]	0.41*	0.82*	1.64*
VENC [m/s]	0.3	1.2	1.7
Echo time [ms]	4.6	3.2	3.1

* The Re and mean flow velocity refers to the un-occluded part of the phantom. In the stenosis center, the Re is doubled and the mean flow velocity is four folded.

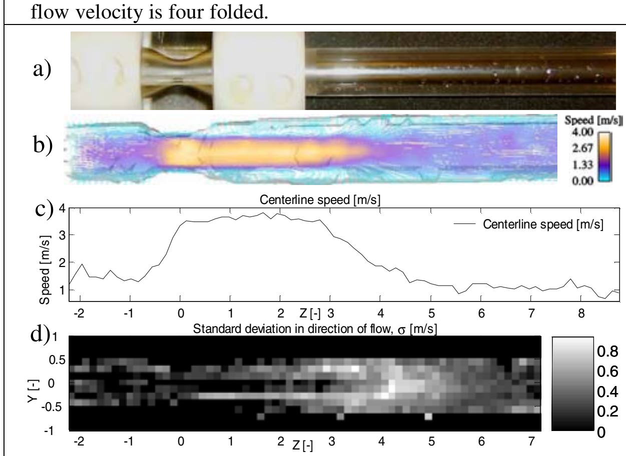

Fig. 1. a) An image of the phantom. b,c,d) Visualization of measurements at Re = 1000. b) A streamline visualization enclosed by an isosurface. c) A plot of the centerline speed. d) A SD map. Y and Z are dimensionless long measures that show the distance from the stenosis center.

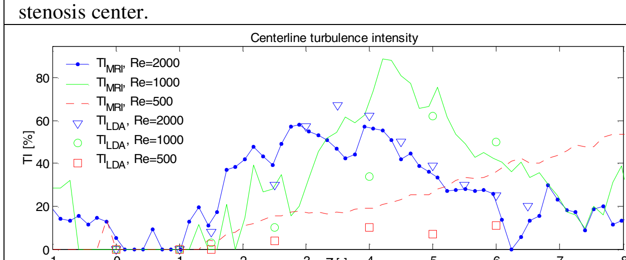

Fig. 2. Centerline TI for Re 500, 1000, 2000 in the flow-direction as measured with MRI and LDA. On the horizontal axis, the dimensionless long measure Z shows the distance from the stenosis center.

Results

Some characteristics of post-stenotic turbulent flow appear in the visualization of the velocity measurement for Re = 1000 (fig. 1b,c). The streamlines in fig. 1b shows a recirculation zone surrounding a flow jet distal to the stenosis and fig. 1c shows an elevated level of centerline speed that suddenly decreases at Z ≈ 3. Fig. 1d depicts low SD in the center of the flow jet and high SD distal to the point where the speed decreases. The comparison with the LDA results is seen in fig. 2.

Discussion

The results show that the presented method has great potential of measuring SD and TI. As seen in fig. 2, the TI measured with our method has the same outline as the LDA results. The displacement of peak turbulence intensity between different Reynolds numbers agrees well with the LDA results.

The method for measuring SD and quantifying TI has the potential of becoming a powerful tool for several applications including assessments of heart valves, detection and analysis of turbulence in predilection areas of atherosclerosis, and quantitative studies of turbulence in industrial flow systems.

References

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