IN VITRO STUDY OF WALL SHEAR STRESS MEASUREMENT USING A NON-INVASIVE ECHO-PIV TECHNIQUE

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INTRODUCTION

Wall shear stress in hemodynamics has significant correlation with amplification or attenuation of a variety of cardiovascular processes. For example, previous studies have shown that wall shear stress is related to the growth of atherosclerosis [1]. To fully understand how shear stress affects such processes wall shear stress must be measured precisely. Most previous studies used computational or *in vitro* measurement techniques to investigate wall shear stress in simulated anatomies and blood flow conditions [2,3]. Although such models can give highly precise results, deviation from the *in vivo* environment always limits the validity of the results. To overcome this shortcoming, accurate *in vivo* measurement of wall shear stress is needed. Some studies have been performed using magnetic resonance imaging (MRI) [4,5], which allows measurement of azimuthal (circumferential) variance in wall shear stress. However, MRI is a cumbersome method, requiring long scan times and poor temporal resolution.

Several researchers have measured wall shear stress *in vitro* precisely and non-intrusively using ultrasound Doppler [6,7]. While ultrasound is attractive due to its ease of use, several limitations including angle dependence, ability to measure only one component of velocity, etc., limit its application. We have developed a new ultrasound-based technique with no angular dependence capable of measuring a time-resolved, two-component, two-dimensional flow field. The technique, known as echo particle image velocimetry (Echo PIV) takes advantage of the non-linear backscatter characteristics of small (2 – 5 μ) gas filled microbubbles seeded into the blood stream in order to perform precise particle detection and velocimetry tracking [8,9]. In this study, we show that this method can be applied to the measurement of wall shear stress.

EXPERIMENTAL APPARATUS AND METHOD

To estimate the accuracy of wall shear stress measurements using echo-PIV, a fully developed laminar pipe flow was used. A schematic of the laminar flow apparatus is shown in Figure 1. A pump delivered flow to a constant head pressure reservoir and the flow rate was controlled by a distal valve. The inner diameter (d) and thickness of the acrylic tube was 0.95 and 0.16 cm, respectively. The mean velocity was 0.11m/s and the Reynolds number based on this mean velocity and pipe diameter was 1002. From this, the entrance length for full development was calculated to be 0.57 m, so a 1.8 m long pipe was used to guarantee fully developed laminar flow condition. A bulk flow meter provided a volume flow rate for estimation of the analytic parabolic velocity profile. Optison□, an ultrasound contrast agent consisting of perflourocarbon-filled microbubbles encapsulated in a lipid-protein sheath, was injected into the constant head reservoir and a stirring device was used in the reservoir to create a uniform distribution of the micro-bubbles.

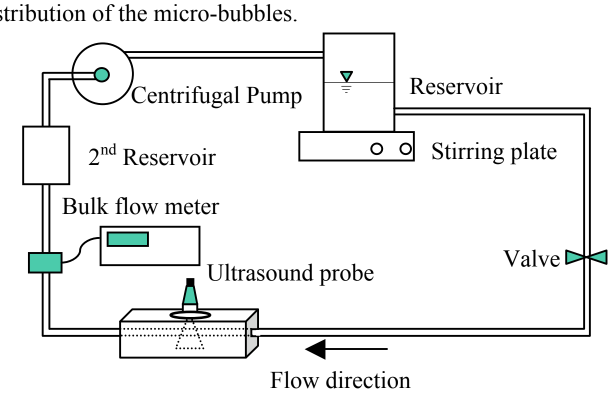

Figure 1. Pipe flow apparatus.

The echo images were obtained using a commercial clinical ultrasound apparatus (GE/Vingmed Vivid 5). A broadband transducer (10 MHz maximum frequency) was used to generate a 6.7 MHz beam. The final interrogation window size was 8□8 pixel with 75% overlapping and the corresponding spatial resolution was 1.2(r) □1.7(□) mm. A more detailed description of the echo-PIV analysis is provided in references [8] and [9].

RESULTS

Figure 2 shows the comparison between the normalized velocity profile obtained from echo PIV and the estimated analytic velocity profile. The echo PIV result is the mean of 50 instantaneous fields.

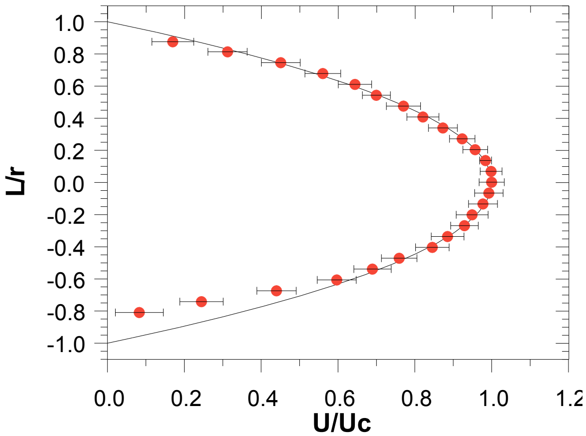

Figure 2. Comparison of echo-PIV and analytic velocity profile (• : Echo-PIV , - : from analytic solution)

Except near the proximal wall boundary region (near field of the ultrasound image), the echo PIV results show very close agreement to the analytic velocity profile. Scattering artifact due to reverberation was consistently observed in the near field; this distorted the velocity vector calculation by including apparently stationary particles in the interrogation windows near L/r = -1.0. The far field (L/r = +1.0) results were used to calculate the wall shear stress.

Figure 3 shows the effect of curve fit choice and overlap ratio on error between the measured and theoretical shear stresses. In this study, we used both 2nd and 3rd degree polynomial curve fits; far better results are obtained from the 3rd order fit. These results confirm Fatemi and Rittgers' work [10], which showed that wall shear stress measurement using direct calculation from the velocity gradient is very sensitive to the curve fitting function. They also demonstrated superior results from a 3rd degree polynomial fit using 4 points; one is 0 velocity from the no-slip condition at the wall and three are measured points. We investigated the effect of using an assumed 0 velocity first point versus using the actual data point nearest the wall, and found that the actual data produced better results.

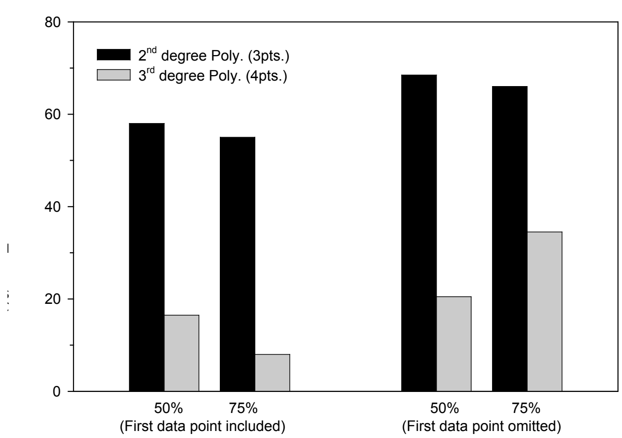

Figure 3. Effect of velocity profile fitting function on shear stress measurement accuracy.

The third effect studied was the distance between adjacent measurements. This was varied by changing the overlap ratio in the PIV algorithm. Increasing the overlap ratio can decrease the distance between two consecutive measurement points, although the actual spatial resolution, determined by the interrogation window size, is not changed. Figure 3 showed that a higher overlap ratio, 75%, coupled with the 3rd degree fit and inclusion of the first data point resulted in the best shear stress measurement, with 8% error.

CONCLUSION

This study demonstrates that echo PIV can be used for non-invasive wall shear stress measurements. This technique uses common clinical methods, and holds promise for *in vivo* measurement of vascular wall fluid shear stress.

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