Ultrasound in Med. & Biol., Vol. ■, No. ■, pp. 1–9, 2016

0301-5629/$ - see front matter

http://dx.doi.org/10.1016/j.ultrasmedbio.2016.06.009

● *Original Contribution*

INTRA-OPERATIVE VECTOR FLOW IMAGING USING ULTRASOUND OF THE ASCENDING AORTA AMONG 40 PATIENTS WITH NORMAL, STENOTIC AND REPLACED AORTIC VALVES

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(*Received* 16 *December* 2015; *revised* 1 *June* 2016; *in final form* 6 *June* 2016)

Abstract—Stenosis of the aortic valve gives rise to more complex blood flows with increased velocities. The angle-independent vector flow ultrasound technique transverse oscillation was employed intra-operatively on the ascending aorta of (I) 20 patients with a healthy aortic valve and 20 patients with aortic stenosis before (IIa) and after (IIb) valve replacement. The results indicate that aortic stenosis increased flow complexity ($p < 0.0001$), induced systolic backflow ($p < 0.003$) and reduced systolic jet width ($p < 0.0001$). After valve replacement, the systolic backflow and jet width were normalized ($p < 0.52$ and $p < 0.22$), but flow complexity was not ($p < 0.0001$). Flow complexity ($p < 0.0001$), systolic jet width ($p < 0.0001$) and systolic backflow ($p < 0.001$) were associated with peak systolic velocity. The study found that aortic stenosis changes blood flow in the ascending aorta and valve replacement corrects some of these changes. Transverse oscillation may be useful for assessment of aortic stenosis and optimization of valve surgery. (E-mail: lindskov@gmail.com) © 2016 World Federation for Ultrasound in Medicine & Biology.

Key Words: **Intraoperative ultrasound, Transverse oscillation, Vector flow imaging, Aortic valve stenosis, Systolic backflow, Secondary flow, Flow complexity.**

INTRODUCTION

Stenosis of the aortic valve is a degenerative process and is one of the most common types of valvular heart disease in Western countries, with older age as an independent determinant, having a prevalence of 4.6% in persons >75 y (Manning 2013; Nkomo et al. 2006). Even though aortic valve stenosis is not considered a major health care problem, the economic costs related to aortic valve stenosis are growing as the general population in Western countries is aging and as a consequence of improved diagnostic methods and new treatment options (Badheka et al. 2015; Nkomo et al. 2006).

The common modality for evaluation of aortic valve stenosis is transthoracic echocardiography, which provides anatomic assessment of the heart and aorta; with the use of Doppler ultrasound (US), a non-invasive examination of aortic blood flow can be performed with, for example, measurements of aortic valve peak velocities, mean gradients, aortic valve area and aortic regurgitation (Manning 2013). Furthermore, flow complexity can be assessed by estimation of spectral broadening in spectral Doppler and power intensity in power Doppler and by evaluation of mosaic patterns using color Doppler (Cloutier et al. 1995; Hutchison et al. 1996; Stringer et al. 1989). However, conventional Doppler US has a major limitation in terms of angle dependency, as only the component of blood velocity directed along the axis of the emitted US beam is measured. This implies that assumption of flow direction is necessary in conventional Doppler for flow quantification and that the flow estimation is limited by the beam-to-flow angle, which should be kept below 70° (Evans et al. 1989).

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The first efforts in solving the angle dependency of conventional Doppler systems were made several decades ago (Bonnefous 1988; Fox 1978; Newhouse et al. 1987; Trahey et al. 1987). Today, a commercial US system by BK Medical (Herlev, Denmark) is equipped with angle-independent vector velocity estimation for blood flow. The method is based on the vector velocity method transverse oscillation (TO) proposed by Jensen and Munk (1998); this method enables real-time and angle-independent blood flow estimation. TO has been tested in computer simulations and with flow phantoms (Udesen and Jensen 2006) and has been validated *in vivo* against magnetic resonance imaging (MRI) angiography (Hansen et al. 2009) and conventional spectral Doppler US (Pedersen et al. 2012). Finally, results on epicardiac intra-operative TO scans of a few patients have been published (Hansen et al. 2015b), as have two recent studies concerning velocity and volume flow quantification (Hansen et al. 2015a) and evaluation of systolic backflow and secondary helical flow (Hansen et al. 2016), in the same 25 patients. The group of patients with normal aortic valves included in this study is taken from the patient population of the two aforementioned articles (Hansen et al. 2015a, 2016).

The aim of the study was to examine the influence of aortic valve stenosis on blood flow in the ascending aorta by comparing blood flow in 20 patients with a healthy aortic valve with that of 20 patients with aortic valve stenosis before and after valve replacement. The hypothesis was that intra-operative US examination with TO is useful for detection of flow differences in the ascending aorta among patients with normal, stenotic and replaced aortic valves.

METHODS

The study was approved by the local ethics committee (No. H-2-2012-039). Twenty patients (15 men, 5 women, mean age: 67.6 y, range: 53–80 y) without a history of valvular disease undergoing coronary bypass surgery and 20 patients (15 males, 5 females, mean age: 74.6 y, range 66–85 y) with aortic valve stenosis undergoing valve replacement with a biological valve entered the study after submitting written informed consent. The 20 patients with normal aortic valves were taken from the patient population of two recent studies of 25 patients (Hansen et al. 2015a, 2016) and were selected by age and gender for best possible match to the patient group with aortic valve stenosis.

Epi-aortic scan sequences of blood flow in the ascending aorta were recorded with TO after standard sternotomy and opening of the pericardium and before cannulation for extracorporeal circulation. For patients with aortic valve stenosis, the epi-aortic scan after replacement with the biological prosthetic valve was performed after decannulation. Standard spectral Doppler measurements of blood flow in the ascending aorta were obtained with transesophageal echocardiography (TEE) before each TO scan.

Transverse oscillation

The TO method uses a conventional pulse for Doppler US, but both the axial and transverse velocity components from each received echo are estimated. The axial velocity is found, as in conventional Doppler US, by using a conventional bell-shaped apodization function in receive. The transverse velocity is found by changing the apodization function in receive to resemble a two-point source and with the use of a special estimator (Jensen and Munk 1998). Combining the axial and transverse velocity components for each point within a region of interest (ROI) provides a 2-D vector velocity map of angle-independent blood velocities. TO has been described in detail (Jensen and Munk 1998; Udesen and Jensen 2006).

A conventional US scanner (ProFocus 2202 UltraView, BK Medical) and a linear transducer (8670, BK Medical) with a center frequency of 9 MHz were used under sterile settings to record epi-aortic scan sequences. To enhance the acoustic transmission from the probe to the aortic surface, warm sterile saline was poured into the mediastinal cavity before each epi-aortic scan (American Society of Anesthesiologists and Society of Cardiovascular Anesthesiologists Task Force on Transesophageal 2010).

The ascending aorta was scanned in long-axis (Fig. 1) and short-axis (Fig. 2) views for each patient. Direction and velocity of the blood flow are given by the color map and indicated by the superimposed *vector arrows*. The temporal resolution of the TO estimation is 16 frames/s, and the maximum scan depth is approximately 5 cm with the transducer setup available. For each scan session, the color box was adjusted to cover the lumen, and the pulse repetition frequency (PRF), depth setting, gain and wall filtering were adjusted for vector velocity estimation. The averaged applied parameter settings for each patient group used in long-and short-axis views are summarized in Table 1. Each scan sequence of 14 s of recording corresponded to 225 frames. On the US scanner, vector velocity estimates are displayed in real time, but without any quantification of velocities available. Thus, scan sequences were analyzed off-line using MATLAB (The MathWorks, Natick, MA, USA), as previously described (Pedersen et al. 2012).

Each scan sequence in the short-axis view was visually inspected for helical motion, and all frames with clear helical motion were analyzed. Using streamlines for displaying the blood motion, the center of the helical