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● *Original Contribution*

ASSESSMENT OF SPECTRAL DOPPLER FOR AN ARRAY-BASED PRECLINICAL ULTRASOUND SCANNER USING A ROTATING PHANTOM

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Abstract—Velocity measurement errors were investigated for an array-based preclinical ultrasound scanner (Vevo 2100, FUJIFILM VisualSonics, Toronto, ON, Canada). Using a small-size rotating phantom made from a tissue-mimicking material, errors in pulse-wave Doppler maximum velocity measurements were observed. The extent of these errors was dependent on the Doppler angle, gate length, gate depth, gate horizontal placement and phantom velocity. Errors were observed to be up to 172% at high beam–target angles. It was found that small gate lengths resulted in larger velocity errors than large gate lengths, a phenomenon that has not previously been reported (*e.g.,* for a beam–target angle of 0°, the error was 27.8% with a 0.2-mm gate length and 5.4% with a 0.98-mm gate length). The error in the velocity measurement with sample volume depth changed depending on the operating frequency of the probe. Some edge effects were observed in the horizontal placement of the sample volume, indicating a change in the array aperture size. The error in the velocity measurements increased with increased phantom velocity, from 22% at 2.4 cm/s to 30% at 26.6 cm/s. To minimise the impact of these errors, an angle-dependent correction factor was derived based on a simple ray model of geometric spectral broadening. Use of this angle-dependent correction factor reduces the maximum velocity measurement errors to <25% in all instances, significantly improving the current estimation of maximum velocity from pulse-wave Doppler ultrasound. (E-mail: da.kenwright@gmail.com) **© 2015 World Federation for Ultrasound in Medicine & Biology.**

Key Words: **Doppler ultrasound, Blood velocity, High-frequency ultrasound, Preclinical ultrasound, Doppler phantom.**

INTRODUCTION

Doppler ultrasound provides a means to measure blood velocity and is used in both research and clinical investigations to quantify the extent and effect of arterial disease. Applications include determination of the degree of stenosis for determining stroke risk (Grant et al. 2003); the downstream resistance to flow to assess renal haemodynamics (Chen et al. 1993); volumetric blood flow, also requiring a measurement of arterial diameter (Alvarez et al. 1993); and wall shear stress as a potential indicator of atherogenic risk (Blake et al. 2008; Yang et al. 2013a). Velocity measurements are typically derived from either the mean or maximum frequency of the Doppler spectrum. The mean frequency is very sensitive to the placement of the sample volume within the flow field, and movement artefacts such as transducer motion or vessel displacements between cardiac cycles can cause the sample volume to move relative to the vessel. In contrast, the maximum frequency of the Doppler spectrum is less likely to change with sample volume placement, and thus, measurements of blood velocity often use the maximum Doppler shift. However, ultrasound systems are susceptible to high measurement errors in maximum blood velocity. For clinical systems, this error has been found to be typically in the range 0%–60%; however, this can increase to >100% when the Doppler angle approaches 80°–90° (Hoskins 1996, 1999). Although the misalignment of the ultrasound beam within the target vessel is corrected for with the angle cursor *via* the Doppler equation, this assumes that the ultrasound beam is received at a single, narrow point on the array (Fig. 1a), when in reality the aperture of a transducer is of a finite width, causing the target velocity vector to subtend a range of angles (Fig. 1b) that are not accounted for (Newhouse et al. 1980). This phenomenon is known as *geometric spectral broadening* and has been reported to be the main source of error in maximum velocity estimation (Hoskins 1999; Hoskins et al. 1999).

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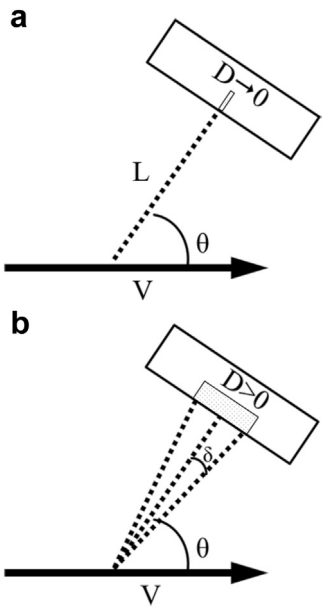

Fig. 1. (a) The Doppler equation, used to calculate the velocity V of a moving target from the Doppler shift in the transmitted and received ultrasound signal, assumes the transducer is at a distance L from the target at an angle to the ultrasound beam θ and does not take into account the size of the aperture D. (b) In reality, the aperture has a finite width ($D > 0$), and therefore, there are a range of angles ($\theta - \delta$ to $\theta + \delta$) that the beam subtends, causing a spread in the received Doppler shift from a target moving with constant velocity.

High-frequency ultrasound is a powerful tool in small animal anatomic and functional *in vivo* imaging because it has high resolution, occurs in real time, is free from ionising radiation and is relatively inexpensive. It is increasingly used as an imaging modality in preclinical investigations, where preclinical relates to models of human disease and often uses small animals such as mice, rats and zebrafish (Goertz et al. 2002; Goessling et al. 2007; Heiss et al. 2008). The first commercially available preclinical imaging system utilised mechanically swept single-element transducers (Foster et al. 2002). A recent study found that for this system, the maximum velocity from spectral Doppler was overestimated by up to 158%, with good agreement with errors predicted from geometric spectral broadening at high beam–target angles (Yang et al. 2013b). As each single-element transducer has a fixed focal depth, the range of useful angles that can be obtained for spectral Doppler measurements is limited. An array-based preclinical system has been developed (Foster et al. 2009). Multiple focal depths can be achieved with electronic focus of the array elements, such that the beam characteristics can be optimised for Doppler measurements over a variety of depths.

Unlike the clinical situation, to date there is limited information on velocity errors in high-frequency ultrasound applications. In this article, we use a small-size rotating phantom to investigate, for the first time, the velocity errors in an array-based preclinical ultrasound scanner, which is predicted to suffer from the same limitations as lower-frequency clinical systems.

METHODS

Ultrasound scanner

Ultrasound scanning was performed using a Vevo 2100 high-frequency ultrasound scanner (FUJIFILM VisualSonics, Toronto, ON, Canada) with a MS-550 D linear-array probe with a central frequency of 40 MHz (broadband frequency: 22–55 MHz). The probe could be set to operate at 32 MHz (default value) or 40 MHz.

Rotating phantom

A miniature rotating phantom composed of tissue-mimicking material (TMM) was created as described by Yang et al. (2013b). The TMM was developed for use with clinical ultrasound systems (Teirlinck et al. 1998) and was recently characterised at high acoustic frequencies (Rajagopal et al. 2015; Sun et al. 2012). Briefly, a cylinder of TMM was set in a mould (inner diameter = 6 mm) on a nylon drive wheel, supported by projecting loop of copper wire. Once the TMM had set, the mould was removed. The drive wheel with the TMM was attached to the drive shaft of the motor of a modified string phantom (BBS Medical Electronics, Hägersten, Sweden). The motor was controllable to provide a constant rotational velocity. The TMM provided ultrasound backscatter such that a Doppler signal could be obtained from the outer edge. The phantom was submerged in a tank filled with 9% glycerol solution by volume, which had an acoustic velocity of 1540 m/s at 20°C. An acoustic absorber pad was placed underneath the phantom to reduce ultrasound reflections. The experimental setup is illustrated in Figure 2a.

Experimental measurements

Pulse-wave (PW) spectral Doppler measurements of velocity were carried out while varying five different parameters: measurement angle; sample gate length; measurement depth (distance from transducer to sample gate); lateral position of the sample gate along the face of the array; and velocity of the rotating phantom. In each case the true linear velocity (V_{true}) was obtained by measuring the period of rotation (T) from a spike in the Doppler trace caused by an indentation in the surface of the TMM, such that

$$V_{true} = \pi d/T \quad (1)$$

where d is the diameter of the TMM cylinder (6 mm). The percentage error in the measured velocity (V_{err}) from Doppler ultrasound ($V_{Doppler}$) was therefore