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● *Original Contribution*

ASSESSMENT OF SPECTRAL DOPPLER IN PRECLINICAL ULTRASOUND USING A SMALL-SIZE ROTATING PHANTOM

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Abstract—Preclinical ultrasound scanners are used to measure blood flow in small animals, but the potential errors in blood velocity measurements have not been quantified. This investigation rectifies this omission through the design and use of phantoms and evaluation of measurement errors for a preclinical ultrasound system (Vevo 770, Visualsonics, Toronto, ON, Canada). A ray model of geometric spectral broadening was used to predict velocity errors. A small-scale rotating phantom, made from tissue-mimicking material, was developed. True and Doppler-measured maximum velocities of the moving targets were compared over a range of angles from 10° to 80°. Results indicate that the maximum velocity was overestimated by up to 158% by spectral Doppler. There was good agreement (<10%) between theoretical velocity errors and measured errors for beam-target angles of 50°–80°. However, for angles of 10°–40°, the agreement was not as good (>50%). The phantom is capable of validating the performance of blood velocity measurement in preclinical ultrasound. (E-mail: ukxinyang@gmail.com) © **2013 World Federation for Ultrasound in Medicine & Biology.**

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

INTRODUCTION

Imaging modalities, including magnetic resonance imaging (MRI) and ultrasound, are finding increasing application in preclinical research (Foster et al. 2000, 2011; Gray et al. 2013; Greco et al. 2012; Moran et al. 2013). These imaging modalities enable longitudinal studies to be performed, increasing the statistical power of investigations with a consequent reduction in the number of animals required. The term *preclinical* generally refers to biomedical research involving the use of small animals, such as mice, rats and increasingly zebra fish, in the development of new diagnostic methods and therapies before trials in humans (*i.e.,* "clinical" research). Key vessels of interest in preclinical work are the aorta, carotid and femoral arteries. These have typical diameters of 0.3–2.0 mm in the rat and 0.15–1.0 mm in the mouse. The typical axial resolution is 50–100 μm for preclinical ultrasound, and in practice, good-quality images of arteries can be obtained in mice and rats. The improvement in spatial resolution of preclinical compared with clinical ultrasound is achieved through the use of higher frequencies. Preclinical ultrasound systems have transmit frequencies in the range 20–50 MHz, compared with 3–12 MHz for clinical ultrasound.

Measurement of blood velocity is performed using the Doppler effect, both in the microcirculation and in major arteries (Christopher et al. 1997; Goertz et al. 2003). Blood velocity has been used as a surrogate for volumetric flow (Bonnin et al. 2008; Hartley et al. 2008; Ishikawa et al. 2003; Li et al. 2008) and for estimation of the degree of stenosis in models of atherosclerosis (Ni et al. 2008).

Although there has been consideration of velocity measurement errors in clinical ultrasound, there is a lack of information on preclinical ultrasound systems. In clinical practice, blood velocity is commonly measured using the maximum Doppler frequency shift. Commercial ultrasound systems overestimate blood velocity, typically by 0%–60%, but this can increase to more than 100% when the Doppler angle approaches 80°–90° (Hoskins 1996, 1999; Hoskins et al. 1991). Typical errors generated in routine clinical practice could

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lead to mis-categorization of patients for carotid surgery (Hoskins 1996).

According to the Doppler equation in its simplest form, a single velocity at any instant in time should give rise to a single Doppler frequency shift at that instant. In practice, a single velocity may give rise to a range of Doppler frequencies. This phenomenon is referred to as *spectral broadening* and may give rise to errors in the blood velocity when estimated from the maximum Doppler frequency. There are a number of sources of spectral broadening (*e.g.,* see Evans and McDicken 2000; Hoskins 2002). The five principal types are:

- *Non-stationarity broadening* is associated with variations in velocity during the sampling time (Fish 1991). This is thought to be relevant mainly during times when the velocity values are changing rapidly, such as in early systole.
- *Velocity gradient broadening* is associated with a range of velocities or directions within the Doppler sample volume. This leads to additional frequencies below the maximum and so, in principle, should not affect maximum Doppler frequency shift.
- *Multi-direction broadening* is associated with a range of velocity directions within the sample volume. This is a significant issue in turbulent flow.
- *Transit time broadening* is associated with the length of time a scatterer remains in the sample volume.
- *Geometric spectral broadening* is associated with the range of angles that the scatterer subtends at the transducer (Censor et al. 1988; Newhouse et al. 1980).

Transit time broadening and geometric spectral broadening had, for a long time, been thought to be equivalent (Newhouse et al. 1980); however, these were shown to be different phenomena by Guidi et al. (2000). For clinical ultrasound systems, it has been found that the maximum Doppler frequency estimation and maximum velocity estimation can be accurately modeled assuming only geometric spectral broadening (Hoskins 1999; Hoskins et al. 1999). This implies that geometric spectral broadening is the main source of error for velocity estimation using clinical ultrasound systems.

In geometric spectral broadening, the finite size of the Doppler aperture means that blood in the sample volume is insonated by a range of angles rather than a single angle. The highest Doppler shift occurs at one extreme edge of the Doppler aperture, whereas in practice, manufacturers perform angle correction with respect to the center of the aperture.

The evaluation of these errors for clinical ultrasound scanners necessitated the development of a range of phantoms involving moving targets, including string and flow phantoms (reviewed in Hoskins 2008). Similar errors are likely to exist for preclinical ultrasound systems, but this is difficult to establish as few Doppler test phantoms have been optimized for preclinical scanners. The aim of this investigation was to develop phantoms for evaluation of Doppler ultrasound-derived velocity values made using preclinical ultrasound systems, with comparison of detected errors with predictions, based on a ray model of geometric spectral broadening.

METHODS

Theory and simulations

The effect of geometric spectral broadening on velocity error was modeled using a previously published ray model (Hoskins 1999). For an un-steered beam produced from a transducer with a width D and focal depth L, the maximum Doppler frequency may be described by the equation

$$\partial F_{max} = (2FV/c)[\cos(\theta)+(D/2L)\sin(\theta)] \quad (1)$$

where F = transmit frequency; V = velocity; c = speed of sound; and θ = beam-target angle.

Equation (1) assumes that the beam width at the focus is zero. For a finite beam width w, we may use the equation

$$\partial F_{max} = (2FV/c)[\cos(\theta)+((D+w)/2L)\sin(\theta)] \quad (2)$$

Typical Doppler systems perform conversion from Doppler frequency to velocity with respect to the center of the Doppler aperture, in which case

$$\partial F_{max} = (2FV/c)\cos(\theta) \quad (3)$$

The error V_{err} in estimated velocity (V_{est}) s defined as

$$V_{err} = (V_{est}-V)/V \quad (4)$$

Rearranging eqns (1), (3) and (4) for the zero-width model yields

$$V_{err} = (D/2L)\tan(\theta) \quad (5)$$

and rearranging eqns (2), (3) and (4) for the finite-width ray model yields

$$V_{err} = [(D+w)/2L]\tan(\theta) \quad (6)$$

For each transducer, eqns (5) and (6) were used to calculate the theoretical error in maximum velocity as a function of angle, using data on aperture size (D), focal depth (L) and beam width (w) provided below.

Ultrasound scanner and beam width measurement

Ultrasound scanning was performed using a Vevo 770 high-frequency ultrasound system (VisualSonics, Toronto, ON, Canada), which has a range of single-element