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● *Original Contribution*

LIMITS OF UNCERTAINTY IN MEASURED VALUES OF EMBOLUS-TO-BLOOD RATIOS IN DUAL-FREQUENCY TCD RECORDINGS DUE TO NONIDENTICAL SAMPLE VOLUME SHAPES

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Abstract—Transcranial Doppler (TCD) ultrasound (US) is widely used to detect cerebral embolisation. A drawback of the technique is that it is difficult to distinguish between small gaseous emboli and large solid emboli. One potential solution is to make measurements from blood and emboli at two different frequencies, because the frequency-dependence of ultrasonic backscatter will be different for the two types of emboli. Unfortunately, it is not possible to produce two identical sample volumes at two different frequencies, and this will introduce an uncertainty into any such measurements. Experimental measurements have been made of US fields distorted by temporal bone at 2.0 MHz and 2.5 MHz, and a mathematical model has been developed to examine the uncertainty introduced by the different field shapes at these frequencies. The results suggest that the uncertainties are sufficiently large (on the order of 2 to 4 dB) that a significant percentage of emboli are likely to be misclassified by the dual-frequency method. (E-mail: dhe@le.ac.uk) © 2005 World Federation for Ultrasound in Medicine & Biology.

Key Words: **Doppler ultrasound, TCD, Beam shape, Emboli, Temporal bone, Measured embolus-to-blood ratio, Dual frequency.**

INTRODUCTION

Transcranial Doppler (TCD) ultrasound (US) is a sensitive technique for the detection of the passage of micro-emboli through the cerebral circulation. The basis of the technique is that if the acoustic properties of emboli are sufficiently different from those of blood, then, when an embolus passes through the Doppler sample volume, there is a transient increase in backscattered power in the Doppler signal that can be detected and analysed. The technique has been described in many publications, and the reader is referred to Angell and Evans (2003) for a recent introduction to the methodology and to Evans (2003) for an overview of the state of the art.

Although detection of emboli is fairly straightforward, determining the composition of an embolus is much more problematical. The magnitude of the Doppler power scattered from an embolus is primarily determined by its composition and its size, with gaseous emboli backscattering very much more power than similarly sized solid emboli. When a very large embolic signal (at 2 MHz, $>$ 35 dB relative to the signal backscattered from blood) is detected, then it is likely that the embolus is gaseous (Evans 2003), but smaller signals may be due either to small gas bubbles or a relatively large solid emboli. It is generally assumed that, given two emboli of the same size, gaseous emboli are likely to be less hazardous because they will dissolve relatively easily in the bloodstream, but even if this were not the case, a large solid embolus giving rise to the same power in the Doppler signal as a small gaseous embolus will certainly be more likely to block blood vessels within the brain and, consequently, cause damage to the brain itself. Because of this difficulty, there has been a considerable research interest in methods for distinguishing between signals from solid and gaseous emboli.

One possible solution is to make measurements of the ratios of backscattered power from the embolus (plus blood) and from blood in the absence of an embolus (*i.e.*, the "measured embolus-to-blood ratio" or MEBR) at more than one frequency. In general, the frequency-dependence of MEBR will be different for small gaseous bubbles and large solid particles over the range of sizes

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where MEBR alone is not sufficient to distinguish between the two types of embolus. For bubbles at frequencies above their resonant frequency, the backscatter cross-section has a constant value irrespective of frequency and, therefore, MEBR (2.0 MHz) is approximately 4 dB greater than MEBR (2.5 MHz) (as the scatter from blood increases with the fourth power of frequency). For very small particles, the backscatter cross-section increases with the fourth power of frequency and, therefore, MEBR (2.0 MHz) is approximately equal to MEBR (2.5 MHz). This relationship begins to break down as particle size approaches the US wavelength because scattering is no longer in the Rayleigh region, and MEBR (2.0 MHz) becomes greater than MEBR (2.5 MHz).

The use of more than one frequency to differentiate between solid and gaseous particles has been explored in some detail, both theoretically and in flow rigs, by Moehring and colleagues (Moehring and Klepper 1994; Moehring and Ritcey 1996; Moehring et al. 1996), and is now available in at least one commercial instrument (Russell and Brucher 2002).

A necessary assumption of the theory of dual-frequency methods for distinguishing gaseous from particulate emboli is that the Doppler sample volumes at both frequencies are identical, or at least very similar. In practice, it is impossible to achieve identical sample volumes at more than one frequency, and the aim of the work described in this paper was to determine how dissimilar field shapes from simple single-element pulsed-wave transducers of the type used in standard TCD machines are likely to be in practical measurement situations and, thus, to ascertain the likely measurement errors and their potential significance. Experimental measurements of beam shapes at the relevant frequencies have been made for the free-field situation, and in the presence of distortion caused by samples of human temporal bone. A mathematical model of the interaction between the experimentally determined beam shapes and idealised segments of the middle cerebral artery (MCA) has been developed, and the likely errors in determining the difference between MEBR at two frequencies have been calculated.

METHODS

The transducer used for all experimental measurements was a DWL (DWL Elektronische Systeme GmbH, Singen, Germany) transducer specifically designed for use with the DWL Embo-Dop® multifrequency US Doppler instrument, intended for discriminating between gaseous and particulate emboli. The diameter of the crystal in the DWL transducer was measured to be 16 mm. The ultrasonic driving frequencies were chosen to be 2.0 MHz and 2.5 MHz, in line with the two operating frequencies of the Embo-Dop® system (Russell and Brucher 2002). The focal length of the transducer was measured in an acoustic tank using a Precision Acoustics (Dorchester, UK) 1-mm needle hydrophone and found to be approximately 50 mm at 2.0 MHz and 55 mm at 2.5 MHz (both of which correspond to a radius of curvature of approximately 80 mm; see Appendix A).

Measurement of beam shapes

Experimental beam-shape measurements were made in an acoustic tank, both without and with temporal bone in front of the transducer. The bone samples had been obtained *post mortem* from the temporal window region of skulls of male cadavers 25, 78, 83 and 84 years old. The samples were approximately 2 × 2 cm in size and were stored in formalin following extraction. These were the same samples used by Angell and Evans (2003).

For measurements without bone, the DWL transducer was coupled directly to the acoustic window of the tank, constructed from 50-μm polyethylene sheeting, with care being taken to remove air bubbles from the gel. For measurements with bone, the samples were placed between the transducer and the tank window, each side being coupled with acoustic transmission gel.

Data acquisition was supervised by a Dell Pentium III-based PC (Dell Inc, Round Rock, TX) that controlled the positioning of the hydrophone within the beam-plotting tank, the pulsing of the transducer and the acquisition of digital data. The beam plotter was controlled through a purpose-built interface that could accurately position the 1-mm diameter Precision Acoustics hydrophone anywhere within the region-of-interest (ROI). The transducer was driven by a HP 33120 signal generator (Agilent Technologies, Palo Alto, CA) that was controlled *via* a GPIB interface (National Instruments GPIB-USB-B; Austin, TX) by software running on the Dell. The hydrophone output was connected to a Precision Acoustics submersible preamplifier and then to a Precision Acoustics amplifier, and a further radiofrequency (RF) amplifier (constructed in-house). Finally, the output of the RF amplifier was connected to an analogue-to-digital converter (ADC). The acquisition hardware consisted of a Signatec PDA 12 high-speed ADC card (Signatec Inc., Corona, CA) connected to the host computer *via* the PCI bus. It was configured to acquire 16,384 data points with 12-bit resolution at 100 MHz. At each acquisition position, a 1498-point section of data (150 μs), positioned so as to span the centre of the required received signal, was saved to disk.

During each experiment, a 3-D array of values of acoustic pressures in front of the transducer was collected in a series of 2-D planes parallel to its face, at axial distances from the centre of the transducer face of be-