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● *Original Contribution*

RELATIVE BLOOD FLOW CHANGES MEASURED USING CALIBRATED FREQUENCY-WEIGHTED DOPPLER POWER AT DIFFERENT HEMATOCRIT LEVELS

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Abstract—In theory, the power of a trans-cranial Doppler signal may be used to measure changes in blood flow and vessel diameter in addition to velocity. In this study, a flow index (FI) of relative changes in blood flow was derived from frequency-weighted Doppler power signals. The FI, plotted against velocity, was calibrated to the zero intercept with absent flow to reduce the effects of non-uniform vessel insonation. An area index was also calculated. FIs were compared with actual flow in four silicone tubes of different diameter at increasing flow rates and increasing hematocrit (Hct) in a closed-loop phantom model. FI values were strongly correlated with actual flow, at constant Hct, but varied substantially with changes in Hct. Percentage changes in area indexes, relative to the 4-mm tube, were strongly correlated with tube cross-sectional area. The implications of these results for *in vivo* use are discussed. (E-mail: seanwallace@live.no)

Key Words: **Trans-cranial Doppler, Flow index, Area index, Doppler power, Hematocrit.**

INTRODUCTION

Trans-cranial Doppler (TCD) is a well-established, non-invasive method commonly used to measure blood flow velocity (BFV) in the major intracranial vessels. However, a major limitation of this method is its inability to measure blood flow directly, because blood flow in a vessel is determined by both BFV and vessel cross-sectional area (CSA). In the clinical setting, information regarding changes in blood flow is inferred from changes in BFV on the assumption that vessel CSA remains constant. Several studies have validated a relationship between BFV and blood flow in specific clinical situations (Batton et al. 1983; Bishop et al. 1986; Greisen et al. 1984; Haaland et al. 1994; Hansen et al. 1983; Larsen et al. 1994; Lindegaard et al. 1987; Trivedi et al. 1997). In most clinical settings, however, this assumption is either incorrect or unproven (Kontos 1989; Sonesson and Herin 1988; Weyland et al. 1994). This potential source of error may be eliminated if some form of vessel diameter measurement is made together with velocity recordings. Such measurements are, however, difficult in cerebral vessels, as the accuracy of imaging techniques is restricted by the small dimensions of the vessels and any diameter changes that occur.

A method that potentially avoids the need for direct measurement of CSA uses changes in the power of the Doppler signal. It has been postulated that each red blood cell (RBC) contributes equally to the power of the reflected Doppler signal, provided the vessel is insonated with uniform intensity. The power of the received signal should therefore be related to the number of RBCs and, hence, the volume of blood within the sample volume (Arts and Roevros 1972).

The reflected signal power is, however, also affected by variations in the spatial positioning of RBCs relative to each other. The random positions of RBCs, or aggregates of RBCs, in the Doppler sample volume produces speckling in the received Doppler power. With non-turbulent flow, because of the large number of RBCs, there is an average distance between the randomly positioned cells at each hematocrit (Hct). The received Doppler power should depend on the Hct, as well as flow velocity and vessel lumen size.

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Because of the specific velocity patterns of blood flow within a vessel, the Doppler signal comprises a spectrum of frequencies, each relating to a specific velocity of red blood cells and their aggregates within the blood flow profile. If the axial length of the sample volume does not change, then the sum of the power levels of these frequencies should be proportional to the volume of blood within the sample volume (Hatab et al. 1997).

Given that volume flow (mL/min) = mean velocity × cross-sectional area

$$F = V_{\text{mean}} \times A \tag{1}$$

a flow index (FI) can be calculated from the sum of the products of Doppler frequency (f_i) and corresponding power signal (P_i):

$$\text{FI} = \sum P_i \times f_i \tag{2}$$

An area index (AI), which can be used to determine relative changes in CSA, may then be calculated by dividing the FI by the mean or maximum velocity of the Doppler spectrum:

$$\text{AI} = \text{FI}/V_{\text{max}}. \tag{3}$$

The correlation between the frequency-weighted power (FWP) and blood flow is based on the power of the insonating beam reflected from within the vessel. FIs are therefore affected by a number of factors, besides speckle, that cause a non-uniform intensity distribution of the ultrasound beam within the vessel (Deverson and Evans 2000a, 2000b; Fry and Barger 1978; Hatab et al. 1997). This, in turn, influences the correlation between FI and V_{max}. The resulting FI/V_{max} correlation line will have a slope that is off-set, so that it does not pass through the zero axis intercept. This line should pass through the zero intercept, as there is no flow when velocity in the vessel is zero. This off-set may, in theory, be corrected by calibration of the FI/V correlation line to the zero intercept, thereby reducing the effects of the aforementioned confounding factors.

The use of FWP to assess relative blood flow changes assumes not only that on average each RBC contributes to the reflected Doppler power, but also that the concentration of RBCs remains constant. The number of RBCs contributing to the reflected signal is dependent on both the volume of blood within the sample volume and the Hct, where Hct is defined as the percentage volume of blood occupied by RBCs. There is a linear relationship between the reflected Doppler signal power and RBC concentration at low Hct values, with maximal power at a Hct of approximately 28% (Mo et al. 1994; Oates 2001; Yuan and Shung 1988). This is lower than physiologic levels, which are normally around 40%. Hct values greater than 28% should also be associated with speckle, and are associated with a decrease in the Doppler signal power caused by clumping of RBCs and multi-reflections of the Doppler signal (Oates 2001; Atkinson and Woodcock 1982).

The aims of this *in vitro* study were, first, to determine the accuracy of flow index and area index measurements with off-set calibration when used to assess relative changes in blood flow and cross-sectional area and, second, to study the effects of hematocrit changes on these measurements.

METHODS

Closed-loop phantom

Studies were performed using a closed-loop system of silicon tubes (RCT High-Flexible, Reichelt, Heidelberg, Germany) containing saline and human whole blood. The blood, which had exceeded its clinical usage date the previous day, was obtained from the local blood transfusion bank. Non-pulsatile, forward flow was generated using a digital roller pump (Ismatec, MCP Process Pump, Glattburg, Switzerland). The blood was heparinized, kept at constant flow and continuously filtered using a 40-μmol micro-filter (BMAF-A arterial filter 40 μm, Medos Medizintechnik, Stolberg, Germany), to prevent contamination by either gas bubbles or solid microparticles. A Windkessel function was incorporated into the system. A constant temperature of 32°C was maintained within the closed-loop system by passing the tubing through a heated water bath. Although lower than normal, this temperature approximates physiologic conditions while providing some protection to the red blood cells during the studies. It has clinical significance, as it is used during cardiac surgery and hypothermic treatment of hypoxic brain injuries. The temperature was continuously monitored using a digital thermometer (Metrawatt M4051, BBC Goerz, Vienna, Austria).

Recordings

Flow index measurements were made at flow rates of 150, 240 and 320 mL/min by insonating silicone tubes with inner diameters similar to those of the middle cerebral artery of children and adults: 1.5, 2, 3 and 4 mm with a wall thickness of 0.5 mm. The tubes were insonated at an angle of 45°, with two-channel TCD instrumentation, 2-MHz insonation frequency, and a peak repetition frequency of 8 kHz (DWL Compumedics, Singen, Germany). The tubes were insonated through a 0.5-cm Plexiglas wall to mimic the effects of the human skull acoustic impedance Plexiglas = 3.2 kg/m^2 s; cortical bone = 3.38 kg/m^2 s (Bloomfield et al. 2000; Smitmans 2002). The transducer was secured with a specially designed Plexiglas holder. Each of the four tubes was