Determination of Coronary Blood Flow Following Coronary Artery Bypass Surgery Using a Bidirectional Doppler System: An Alternative

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Introduction

Consistency and accuracy of electromagnetic blood flow instrument measurements has been questioned at the Barnes Hospital in St. Louis. These instruments are utilized to measure blood flow in reversed autologous saphenous veins during coronary artery bypass surgery. The proper and accurate use of these instruments require that certain conditions must be met: 1) proper matching of the size of the flow transducer to the size of the graft, 2) proper application of the transducer to the vein graft so that a tight fit results in a 20–40% constriction, 3) removal of all extraneous tissue from the site of measurement, 4) minimal magnetic interference, 5) appropriate common grounding, 6) very clean transducer electrodes, and 7) consistent performance of the electronic zero circuitry. The lack of consistency of measurements and the inability in an operating room environment to satisfy all of the above conditions led to a search for alternative systems for measurement of blood flow in coronary artery bypass grafts. Accordingly, a bidirectional Doppler system was evaluated.* Doppler systems measure blood flow velocity as a function of change in sonic frequency and can be used to calculate mean blood flow.

* Medasonics, Inc., Model D9, Mountain View, California 94042.

The purpose of this study was to determine the accuracy, consistency and practicality of a bidirectional Doppler system and to compare this instrument to an electromagnetic flow system. A second evaluation was undertaken to determine the usefulness of the auditory component of the shifted reflected frequencies which qualitatively determine bypass graft patency and estimate average blood flow.

Instrumentation

The Medasonics bidirectional Doppler system is a continuous wave ultrasonic system that uses any one of four frequencies when coupled with matching transducers to measure velocity and direction (2.5, 5.0, 8.0, and 10.0 MHz.) (Figure 1). The unit contains a circuit to produce sound as a function of the forward and reverse frequency shifts (changes in pitch). The output of the amplifier unit is connected to a two channel strip chart recorder to produce analog wave forms of the frequency shifts simultaneously in both the forward and reverse directions.

Principle of Operation

Each Doppler transducer tip contains two piezoelectric crystals protected by a plastic cover, Figure 2. One crystal emits a signal which travels through the

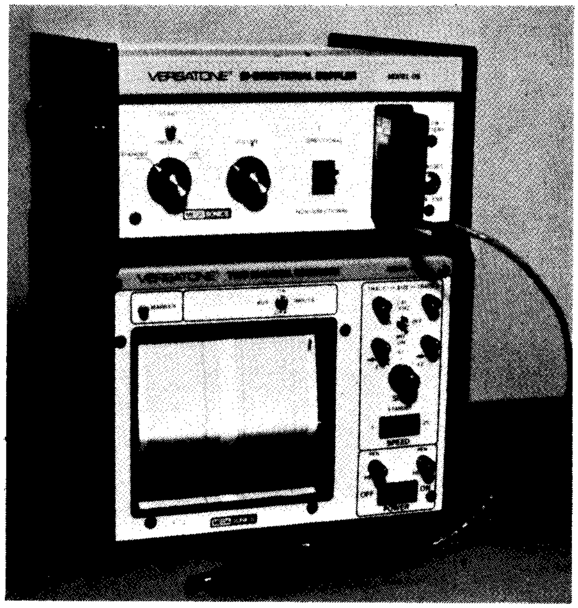

FIGURE 1. The Medsonics Bidirectional Doppler system incorporating an amplifier on top with a 10 MHz pencil probe attached, and two-channel strip chart recorder below.

tissues and the other crystal receives the reflected ultrasound. Constantly emitted frequency signals shift to a higher frequency when the blood flow is toward the probe as the signal strikes a moving mass, i.e., blood flowing in a vessel. A lower frequency shift occurs if the flow is away from the transducer. These shifted frequencies of the reflected signal are transmitted to the Doppler unit where the frequency shift is determined. The frequency shift is amplified and may be heard through a speaker or ear phone or displayed as a wave form (Fig. 3). A sound transmitting gel is used to assure proper sonic contact between the crystals and the tissue. Emitted and received signals will be significantly attenuated if the transducer does not make proper contact with the tissue. Different frequencies are available for monitoring blood flow at different depths. A 10 MHz Transducer is effective for measuring blood velocity at depths up to 2 cm. A 2.5 MHz. transducer will operate efficiently for deeper vessel positions.

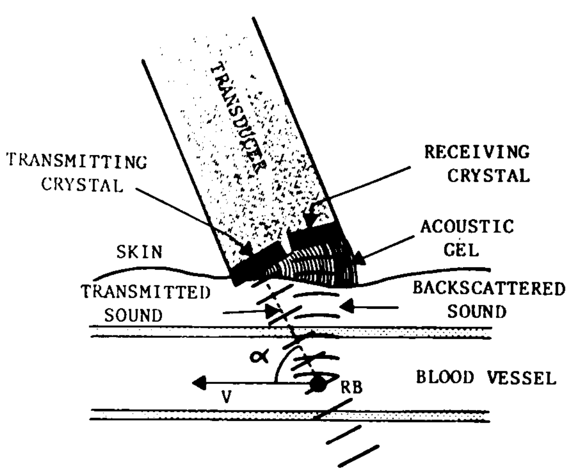

FIGURE 2. A schematic of non-invasive use of the Doppler transducer. α is the angle of the emitted signal with the long axis of the vessel ν is velocity. RB is red blood cells.

Calculations of Blood Flow

Initially, an estimation of the mean frequency of the wave form was necessary to determine mean blood flows.

Mean frequency (Hz)
$= \frac{1}{3}$ [Systolic freq. + (2 × diastolic freq.)].

Mean frequency is used to determine the mean flow velocity by the following equation:

Mean velocity (meters/sec) =

$$\frac{\text{mean shifted freq. (Hz)} \times \text{velocity of sound blood\#}}{2 \cos\theta \times \text{transducer freq. (Hz)*}}$$

Where: θ = the angle between the emitting signal and the vessel wall
the velocity of sound is equal to 1.57×10^3 meters/sec
* the transducer freq. = 10×10^6 Hz.
$\theta = 60°$ ($\cos 60° = 0.5$)

Units of velocity are converted from meters per second to centimeters per minute as follows:

$$\frac{\text{meters}}{\text{sec.}} \times \frac{100\text{ cm}}{1\text{ meter}} \times \frac{60\text{ sec.}}{1\text{ min.}} = 6 \times 10^3\,\frac{\text{centimeters}}{\text{minute}}$$

The angle between the transducer and the vessel wall

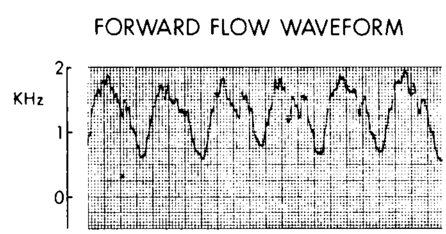

FIGURE 3. Forward blood flow frequency shift wave form measured in a patient with a completed coronary artery bypass vein graft.

is critical to the determination of velocity. A method of maintaining a constant angle(θ) is necessary. A short piece of $\frac{1}{4}''$ I.D. plastic tubing cut at a 60° angle to the long axis of the tubing is utilized. The tubing segment is placed over the tip of the transducer so that a constant 60° angle is created when the transducer is placed against the vessel wall (Fig. 4). An angle of 30 to 60 degrees between the transducer and the vessel wall is necessary to obtain a consistent noise free frequency wave form. A 60° angle is convenient because the cosine of 60° is equal to 0.5, making calculations easier.

Once the mean velocity is calculated, it is necessary to determine the internal cross sectional area of the vessel. This is accomplished while the veins are prepared for grafting (following removal from the leg). The internal diameter of the vessel at a constant internal pressure of 100 mmHg. Next, the vessel obturators of varying size, 1–5 mm., are placed in the smallest end of the vein segment to produce the same outer diameter as that created when the same section of the vein is under intraluminal pressure. A ratio of the outer diameter to inner diameter is used to determine the internal diameter of the graft following implantation and perfusion.

$$\frac{\text{measured vein O.D.}}{\text{measured vein I.D.}} = \frac{\text{anastomosed vein O.D.}}{X}$$

Where: X = the internal diameter of the graft following implant and perfusion.

The internal cross sectional area of the vein is calculated by the formula $A = \frac{1}{4}(\pi \times \text{diameter}^2)$

Mean blood flow is calculated by multiplying the mean velocity by the cross sectional area to yield blood flow in units of ml per minute.

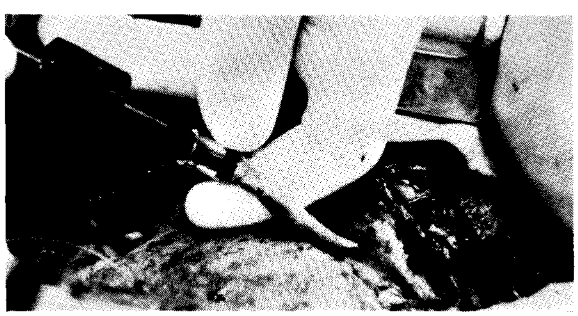

FIGURE 4. Application of the Doppler transducer to a coronary vein bypass graft.

Ex Vivo Testing

Extra portions of saphenous veins are used. The internal and external diameters are measured and the liminal cross sectional areas are calculated. The vein is then attached to a circuit consisting of $\frac{1}{4}''$ I.D. PVC tubing, a pulsatile pump,* and a reservoir (Fig. 5). The vein is attached in parallel with a primary recirculation line. A screw clamp is placed proximal to the vein segment to control the amount of flow. Trials are performed at different flow rates in the range of expected clinical flows (0–160 ml./min.). The tubing is also clamped proximally and distally to the vein segment to study the effects of upstream and downstream occlusion, respectively. Notations are made of the relationship of the measured velocity to the auditory characteristics of the forward and reverse wave forms amplitudes. Additionally, *ex vivo* comparisons of the flow measurements are made between the Doppler system, an electromagnetic flow system† and the actual flows measured with a graduated cylinder and a stop watch. Actual and calculated Doppler flows are determined from three separate timed samples which are averaged for each trial. Finally, comparisons of simultaneous flows in reversed autologous vein grafts in two patients are made between the Dopplers and the electromagnetic flow systems. Augmentation of coronary artery bypass graft flows are examined following the initiation of intraaortic balloon pumping in an additional patient.

Results

A chart which reduces the number of clinical calculations is developed to quickly determine the flow results. Once the mean frequency shift is determined, it is multiplied by a conversion number calculated for selected vessel diameters ranging from 1 to 8 mm. An averaging circuit has also been added to the output of the Doppler system to avoid the need to calculate the mean frequency shift from the recorded phasic wave form.

The results of the Doppler flow determinations as calculated from the velocities and a listing of the actual measured blood flows are shown in Table I. The average difference between the Doppler flow and the actual

* Biomedicus, Minnetonka, Minnesota 55343.

† Biotronex Laboratory, Inc., Model 613, Silver Springs, Maryland 20910.

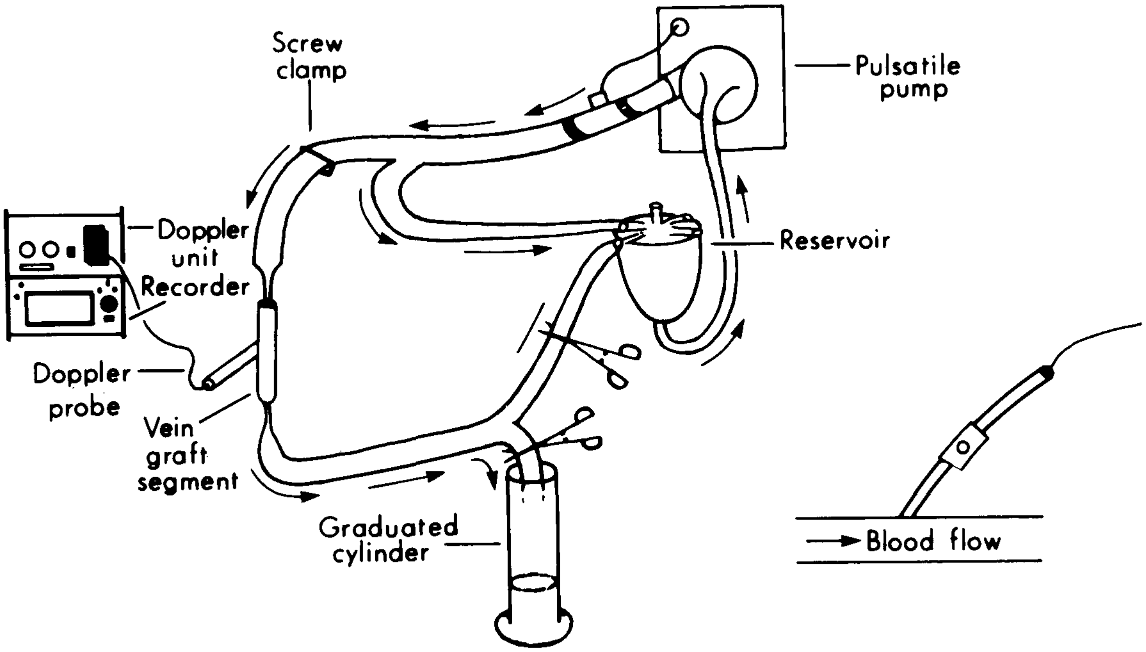

FIGURE 5. The *ex vivo* circuit used for the comparative studies of the Doppler, electromagnetic and actual flows using a pulsatile pump.

flow was 3.25% in ten trials. *Ex vivo* Comparisons of the Doppler, electromagnetic (E-M) and measured flow results are listed in Table II. A 4.5% difference has been found between the Doppler results and the electromagnetic results, while a 6.3% difference was found between those results obtained with an electromagnetic flow system and the actual measured flows.

Our findings show no sound or measurable wave form is evident when the tubing is occluded upstream of the transducer (Fig. 6). When the tubing is clamped downstream relative to the transducer, a low frequency sound is generated and the recorded forward and reversed frequency shifts are low and almost equal (Fig. 7).

TABLE I
Ex Vivo Results of Calculated Doppler Flow Measurements vs. Actual Flow Measurements

Trial	Calculated Doppler Flow (ml./min.)*	Actual Flow (ml./min.)*
1	102	91
2	41	40
3	159	153
4	143	139
5	33	32
6	34	32
7	29	25
8	69	70
9	77	80
10	90	95

* The flows for each trial are averaged from three sample runs timed for one minute each.

The average % difference between the doppler flows and the actual flows = 3.25%

TABLE II
Ex Vivo Results of Calculated Doppler Flow Measurements vs. Electromagnetic Flow Measurements vs. Actual Flow Measurements

Trial*	Calibrated Doppler Flow Measurement (ml. min.)	Electro-magnetic Flow Measurement (ml./min.)	Actual Flow (ml./min.)
1	29	30	25
2	69	65	70
3	77	71	80
4	90	88	95

* Flows for each trial are averaged from three sample runs timed for one minute each.

The average % difference between the doppler and E-M systems = 4.5%.

The average % difference between the E-M system and actual flow = 6.3%.

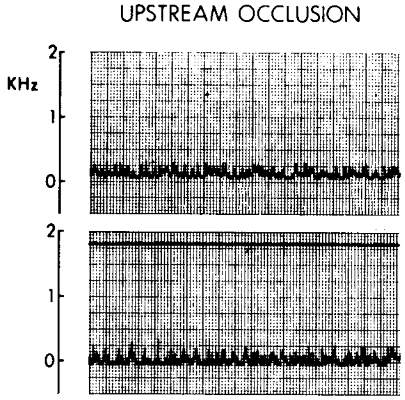

FIGURE 6. Wave forms obtained by occlusion upstream of a vein segment in the *ex vivo* circuit. The upper panel is the forward velocity signal and the lower panel is the reverse velocity signal.

TABLE III
Clinical Results of the Calculated Doppler Flows vs. the E-M Measured Flows

	Calculated Doppler Flows (ml./min.)	E-M Measured Flows (ml./min.)	Vein Graft Site
Patient A	60	80	Right Cor. Artery
	60	65	Left Ant. Descending
	83	55	Diagonal
Patient B	71	80	Posterior Descending
	51	55	Left Ant. Descending
	40	38	Sequential

The % difference in flow results between the doppler and E-M systems = 2.3%

Clinical application of the Doppler system for measurement of blood flow in vein grafts causes no inconvenience to the surgeon. The angled tube over the transducer tip provides a consistent method of producing the same angle for all measurements. The clinical results of six trials comparing the Doppler calculations and the electromagnetic flow measurements can be seen in Table III. A 2.3% difference is found in the results between flow measurements made with the Doppler system and those made with the electromagnetic system.

The intraaortic balloon pump was used in one of our patients and the bypass vein graft flows were measured before and after the initiation of balloon pumping, Table IV. A 21% increase in graft flow (approx.) was noted once the intraaortic balloon pump was started (Fig. 8).

Blood flow in vein grafts has been measured in more than 200 patients with the Doppler system. The calculated flows range from 0 to 155 ml./min. An extensive comparison of the Doppler and electromagnetic systems could not be performed due to the frequent inconsistency of those results obtained with the electromagnetic system.

Discussion

Examination of the *ex vivo* and the clinical data indicate that the Doppler system is an accurate tool and a preferable alternative for the determination of vein graft blood flow during coronary artery bypass surgery. The results in the *ex vivo* trials indicate that the techniques and calculations used to obtain the resultant flows with the Doppler system are acceptable. The

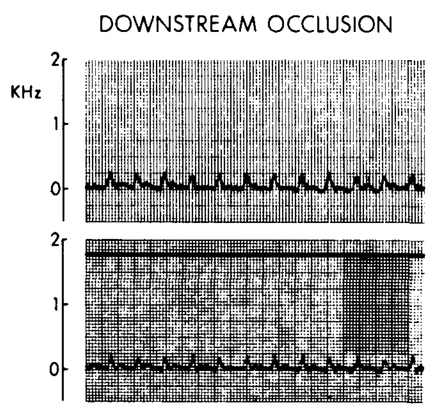

FIGURE 7. Wave form obtained by occlusion downstream of a vein segment in the *ex vivo* circuit. The upper panel is the forward velocity signal and the lower panel is the reverse velocity signal.

TABLE IV
Flow Changes in the Coronary Grafts With and Without Intra-aortic Balloon Augmentation—Calculated with the Doppler System

Vein Graft Site	Calculated Flow with I-A balloon off (ml./min.)	Calculated Flow with I-A balloon on (ml./min.)
Right Cor. Artery	50	60
Left Ant. Descending	43	55
Marginal	33	45

The average increase in coronary graft flow was approximately 21%.

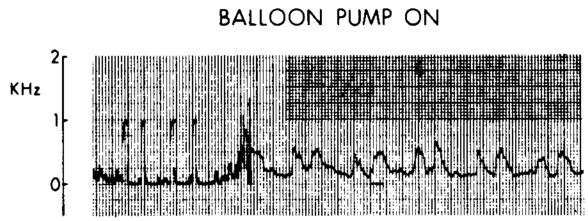

FIGURE 8. The forward velocity wave form obtained before and after intraaortic balloon pumping demonstrating augmentation of vein graft blood flow.

Doppler system is consistently accurate over a wide range of simulated flows ranging from 30 to 160 ml./min. during the *ex vivo* tests.

The clinical results with the Doppler system were in excellent agreement with those obtained simultaneously with the electromagnetic unit. Additionally, the effect of intraaortic balloon pumping on vein graft blood flow was easily determined.

The cost of the Doppler unit, strip chart recorder, and a set of four Doppler transducers is equivalent to a similar electromagnetic flow meter system. The Doppler transducers are sterilized by ethylene oxide and can be aerated briefly before use. The Doppler system is practical and simple to use in the operating room by both the perfusionist and the surgeon.

The results indicate that the Doppler system (in conjunction with the calculation and usage techniques) reported here can be considered a significant cost effective alternative for the measurement of blood flow in bypass vein grafts.

The ability to hear the sound of the blood flow through the vein grafts has given the cardiothoracic team an added diagnostic tool for graft evaluation. The sound of the flow also gives the surgeon the auditory assurance not only of graft patency, but also of the relative velocities in the vein grafts.

Future applications of the Doppler system may include: assessment of the coronary bed run-off, assessment of efficacy of sequential bypass vein grafts, assessment of distal leg perfusion following implantation of an intraaortic balloon, and the measurement of flow redistribution following various pediatric palliative procedures, i.e. pulmonary bandings and shunts.

The Doppler system has been shown to be a useful diagnostic tool in noninvasive applications. This report demonstrates that the Doppler system may be used safely and with accuracy and consistency in the operating room setting.

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