

●Original Contribution

THE TIME-DISTANCE RECORDER AS A MEANS OF IMPROVING THE ACCURACY OF FETAL BLOOD FLOW MEASUREMENTS

P. C. STRUYK,† L. PIJERS,‡ J. W. WLADIMIROFF,‡ F. K. LOTGERING,‡
M. TONGE,‡ and N. BOM§

†Central Research Laboratory, ‡Department of Obstetrics and Gynaecology, §Department of Cardiovascular Research, Erasmus University, Rotterdam, The Netherlands

(Received 15 December 1983; in final form 7 June 1984)

The influence of pulsatile diameter changes on calculation of volume flow has been studied. *In vitro* studies and an animal study were carried out with a real-time imaging and pulsed Doppler velocity measurement system. For precise pulsatile diameter information a wall motion tracking device was incorporated. Whereas *in vitro* a high degree of accuracy was found for the measurements of volume flow, this could not be substantiated in the descending aorta of the fetal lamb, in which Doppler volume flow differed between -7.5 and 17% from magnetic volume flow. In a clinical study the relative influence of various diameter approximations on calculated fetal aortic volume flow was assessed in 16 normal third trimester pregnancies. Depending on the selected diameter approximation method it appeared that differences from 19% underestimation to 9% overestimation in calculated volume flow could be obtained when reference was made to volume flow derived from diameter and velocity information.

Key Words: Pulsed Doppler ultrasound, Time-distance recorder, Fetal blood flow.

1. INTRODUCTION

Since the introduction of a combined 2-D real-time and pulsed Doppler technique for measuring blood flow in the abdominal part of the umbilical vein (Gill, 1979a) and in the fetal descending aorta (Eik-Nes *et al.*, 1980), a number of reports has appeared on fetal blood flow under physiological and pathophysiological circumstances. For a comprehensive review of the current status in this area of fetal research, the reader is referred to the fourth issue of Vol. 10 of this Journal.

Aortic diameter measurements have been carried out from 2-D real-time images (Eik-Nes *et al.*, 1980; Wladimiroff and McGhie, 1981; Griffin *et al.*, 1983; Jouppila *et al.*, 1983) and *M*-mode recordings (Eik-Nes *et al.*, 1982). The pulsatility in aortic diameter is a major source of error in volume flow measurement. Eik-Nes *et al.* (1981, 1982, 1984) have repeatedly pointed out the significance of a time-distance (TD) recorder in registering the changes in the diameter of the pulsating fetal aorta. For correct calculation of flow, knowledge of instantaneous average velocity over the vessel area and the instantaneous vessel area must be known. For the area estimation, in practice an approximation is used based on the time averaged

diameter or a parameter derived from values for the maximum diameter during systole and the minimum diameter during diastole. The area is then calculated assuming rotational symmetry.

The purpose of the present paper was to assess:

(a) The calculated volume flow as obtained with a combined 2-D real-time and Doppler technique together with a TD recording method against the volume flow obtained with an electromagnetic measurement. This was carried out under *in vitro* and *in vivo* animal experimental circumstances.

(b) The error in flow estimation when a comparison was made between various diameter approximations and the diameter as derived from an instrument which did yield diameter as a function of time (TD-method).

2. MATERIAL AND METHODS

A combined dynamically focused linear array 2-D real-time (Organon Teknika) and a 2 MHz pulsed Doppler system (PEDOF) as first described by Eik-Nes *et al.* (1980) were used. The Doppler instrument is capable of estimating the mean frequency shift from the Doppler power spectrum. The mean velocity

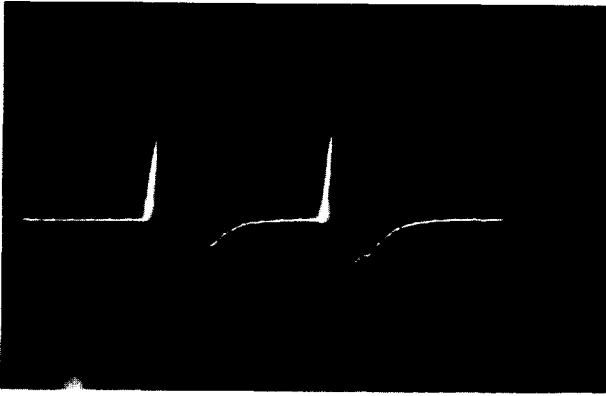


Fig. 1. A-mode presentation of the outer and inner wall of an elastic latex tube. The two bright spots represent the selected zero-crossing tracking points.

over the cross-section of the tube (or the vessel) is derived from the mean frequency if the tube area is completely within the sample volume. A pulsed Doppler transducer with a diameter of 12 mm was used. The sample volume has a length along the ultrasonic beam of 8 mm and a width which is determined by the width of the beam. After the dispersion point, the beam width diverges from 12 to 21 mm at a depth of 11 cm. Uniform scattering is feasible in vessels with a maximum diameter of 5.4 mm at a depth of 7 cm up to a maximum diameter of 9.2 mm at a depth of 10.5 cm. From the 2-D real time image a line was selected for diameter measurement. The basic principle of the time-distance method was described by Hokanson *et al.* (1972). We developed an instrument (Blom, Research Department, Rotterdam) which was capable of following two pre-

selected echoes by a tracking method. From these data an analog output is derived which is proportional to the distance between them. The markers of the TD recorder were positioned on the onset of the deflections of an A-mode representation of both vessel walls (Fig. 1). Thus the changes of the vessel diameter were directly and continuously recorded. Diameter and flow velocity measurements were done at the same level of the aorta.

2.1. In vitro experiments

A pump system (Fig. 2) pulsating at a rate of 140 bpm was used, allowing volume flows to vary between 100 and 700 ml min⁻¹. The circulating fluid consisted of a 0.9% saline solution, in which minute starch particles were dissolved, acting as reflectors for the emitted Doppler ultrasound waves. Mean volume flow (ml min⁻¹) was subsequently calculated in a stiff polyvinyl chloride (PVC) tube and in an elastic latex tube. Magnetic mean volume flow measurements were carried out employing a cannula type magnetic flow probe (type FF series; NIHON/KOHDEN) of 10 mm internal diameter. The error of the magnetic flow probe, which was established before the experiment was less than 5%, which was within the factory specified range. Doppler volume flow in the PVC tube was calculated from the mean flow velocity and internal diameter of the tube. Seven recordings were made from stepwise increased volume flows between 100 and 700 ml min⁻¹. From each recording five consecutive stroke periods were analyzed. Similar recordings were made from the pulsatile flow through the elastic latex tube. This resulted in pulsatile diameter changes as shown in Fig. 3. It may be noticed from Fig. 3 that the peak magnetic flow occurs later

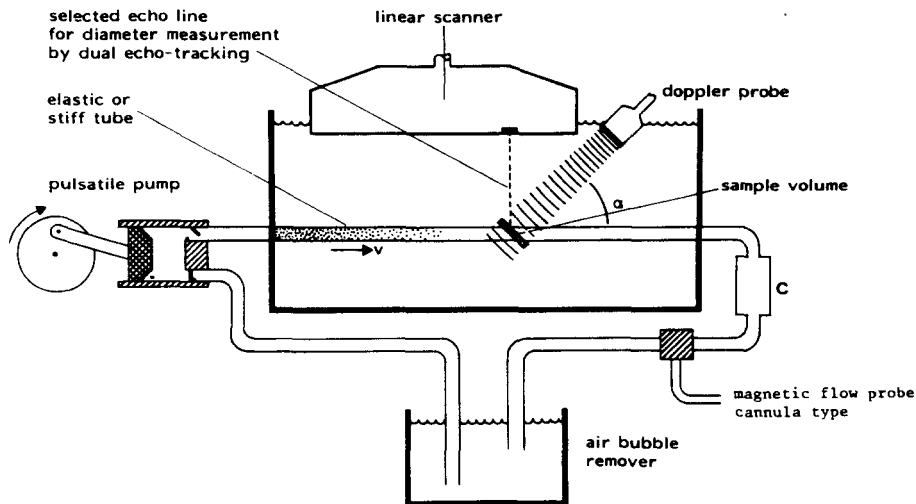


Fig. 2. Set-up for *in vitro* flow tests: V , flow velocity; α , angle of 45° between Doppler beam and flow direction; C , compliance.

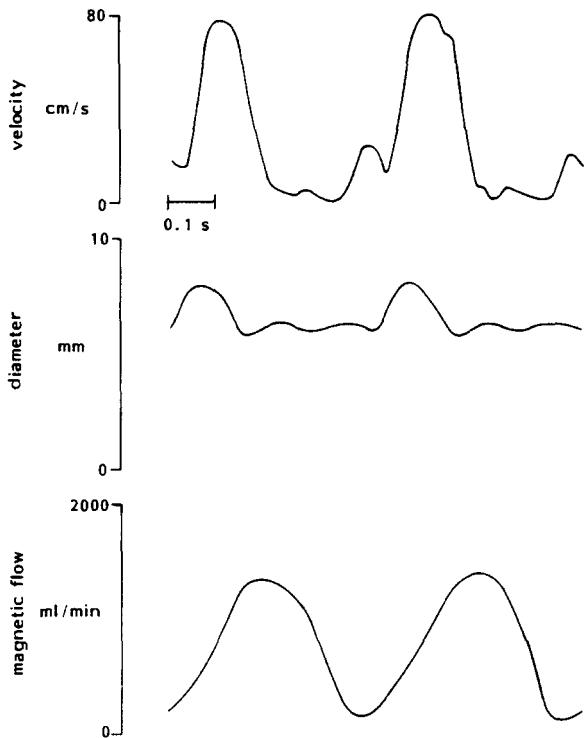


Fig. 3. Flow velocity, tube diameter and magnetic flow in an elastic latex tube.

than the peak Doppler velocity. This is explained by the fact that the magnetic flow is measured after the volume compliance in our experimental set-up. For a pulsatile flow in an elastic pulsating tube, the flow is determined by

$$\bar{F} = \frac{\pi}{4T} \int_0^T v(t)d^2(t) dt$$

where

\bar{F} is the mean volume flow over T , T the integration time, $v(t)$ the flow velocity as a function of time and $d(t)$ the diameter as a function of time.

We sampled both the Doppler flow velocity and the pulsatile diameter signal at a frequency of 33 Hz, corresponding with 15 sampling points during each stroke period (Fig. 4). From the volume flow values calculated at each of the 15 sampling points in each stroke period, the mean volume flow for that particular stroke period was established. The interference between the ultrasound signals from the real-time and Doppler transducer did not allow simultaneous recording of blood flow velocity and pulsatile diameter. Instead, both profiles were recorded immediately after each other. The correct time relationship between the two profiles was obtained by using the magnetic flow profile which was recorded simultaneously with both diameter and velocity profiles.

2.2. Sheep experiment

An acute experiment was performed in a pregnant ewe of 139 days gestation, using standard anaesthesia techniques. With the ewe placed in the supine position, the uterus was exposed through a low midline incision. The uterine wall, membranes and fetal abdomen were opened, and the fetal aorta was dissected from its surroundings. A magnetic perivascular flow probe (Transflow 601 system; Scalar) of 5.5 mm in diameter was placed around the descending part of the fetal aorta, about 1 cm above the origin of the renal artery, and the incisions were closed. The accuracy of the electromagnetic flowmeter used in the sheep experiment was measured in a laboratory set-up. The error was less than 5%.

Blood flow velocity and pulsatile diameter of the descending aorta were documented about 1 cm above the level of the magnetic flow probe employing the same 2-D real-time and pulsed Doppler set-up as was used during the *in vitro* experiment. A maximum time interval of 30 s was allowed between the blood flow velocity and pulsatile diameter recording. Assessment of the correct time relationship between the two profiles was similar to that described in the *in vitro* experiment. Volume flow was calculated from ten equidistant sampling points in each cardiac cycle.

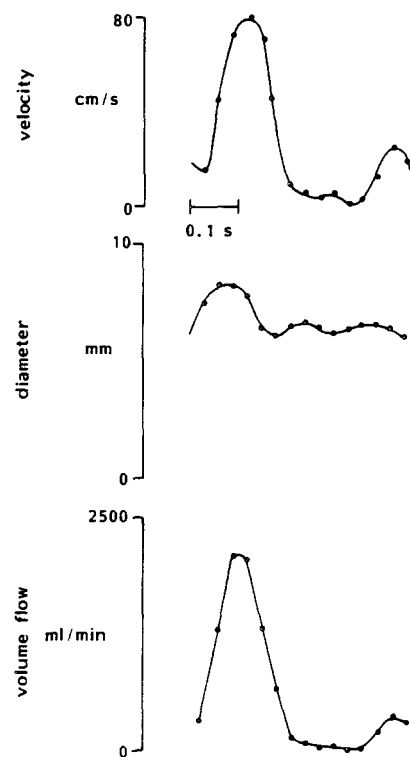


Fig. 4. Volume flow profile in an elastic tube as calculated from the sampled flow velocity and diameter profile.

Fetal heart rate was derived from the flow velocity recordings.

2.3. Clinical study

Blood flow velocity and pulsatile vessel diameter changes were recorded at the lower thoracic level of the fetal descending aorta in 16 normal pregnancies between 30 and 38 weeks of gestation (median 35 weeks). Blood flow was studied in those cardiac cycles during which flow velocity and vessel diameter characteristics were comparable on the basis of equal period times of two consecutive cardiac cycles. A volume flow profile was subsequently constructed from ten equidistant points in the flow velocity and pulsatile vessel diameter profile per cardiac cycle (Tonge *et al.*, 1983). In order to establish the influence of the pulsatile aortic diameter on the calculation of volume flow, the following vessel diameter approximations were considered:

- (a) the maximum vessel diameter within one cardiac cycle;
- (b) the minimum vessel diameter within one cardiac cycle;
- (c) the mean of the maximum and minimum vessel diameter within one cardiac cycle;
- (d) the time-averaged vessel diameter, which is derived from ten sampling points of the pulsatile vessel diameter profile within one cardiac cycle.

These results were compared with a reference or 'effective diameter'. This diameter is derived from

the effective vessel area which is calculated by dividing the time-averaged volume flow by the time-averaged flow velocity. This diameter takes into account the pulsatile nature of the vessel wall as recorded with the TD system.

3. RESULTS

3.1. In vitro experiments

In the stiff PVC tube mean flow velocity was varied between 7 and 44 cm s^{-1} . When comparing Doppler and magnetic volume flow in the stiff PVC tube, a regression line ($y = 0.98x + 29$) close to the line of identity was found ($r = 0.99$; Fig. 5). In the elastic tube mean flow velocity was varied between 7 and 36 cm s^{-1} , the diameter change ranged between 17 and 43% from the diastolic value. When comparing Doppler and magnetic volume flow, whereby the Doppler flow was calculated as described in Section 2 in the *in vitro* experiment, we observed a regression line of $y = 1.16x - 28$ (Fig. 6, regression line I). Apart from an intercept of -28 ml min^{-1} , this results in a 16% overestimate. However, corrected for known wall thickness, apart from an intercept of -28 ml , no overestimation of flow could be demonstrated (Fig. 6, regression line II).

3.2. Sheep experiment

In the sheep experiment fetal heart rate varied between 225 and 235 bpm, while magnetic volume flow ranged between 448 and 520 ml min^{-1} . Doppler

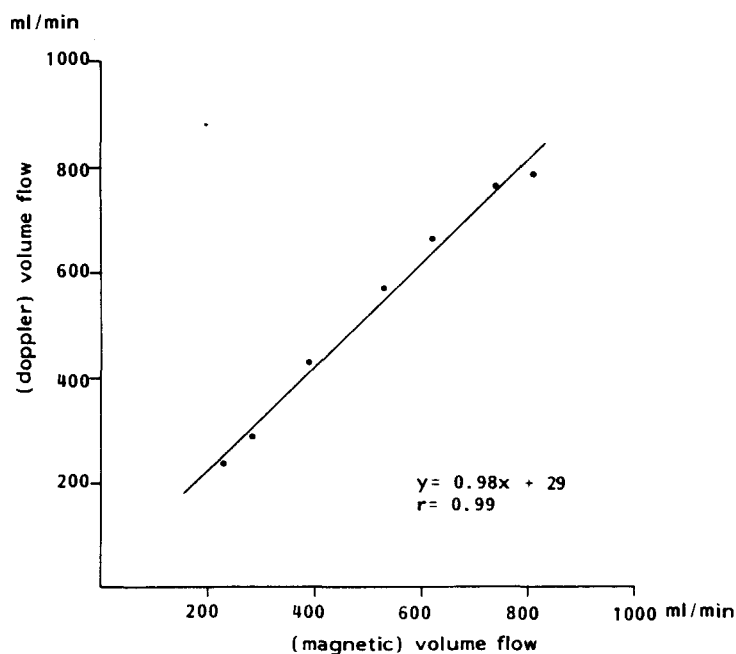


Fig. 5. Relation between Doppler and magnetic volume flow in stiff PVC tube.

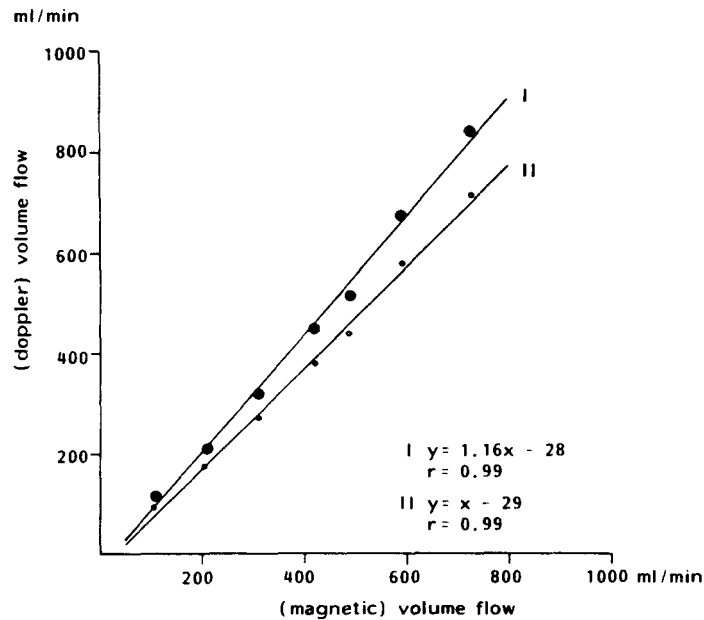


Fig. 6. Relation between Doppler and magnetic volume flow in elastic tube. Line I, Doppler volume flow calculated from the outer-to-inner tube diameter. Line II, Doppler volume flow calculated from the inner-to-inner tube diameter.

flow velocity and pulsatile diameter recordings were carried out at three different periods each lasting 30–60 s. The mean flow velocity varied between 31 and 35 cm s^{-1} and the diameter change ranged between 5.4 and 7.4% from the diastolic value. From each period, ten consecutive cardiac cycles were selected, depicting technically acceptable Doppler flow velocity, pulsatile diameter and magnetic volume flow recordings. In Fig. 7 the calculated Doppler volume flow values are superimposed on the magnetic flow profiles, indicating a good agreement between the two flow profiles. Nonetheless, the differences in volume flow as averaged over ten cardiac cycles between Doppler and magnetic volume flow values were -7.5% (at 520 ml min^{-1}), 17% (at 464 ml min^{-1}) and 5% (at 448 ml min^{-1}).

3.3. Clinical study

The results of the clinical study are given in Table 1. This table demonstrates the differences in volume flow calculated from the maximum diameter, minimum diameter, mean of maximum and minimum diameter and time-averaged diameter when related to the effective diameter.

4. DISCUSSION

The high degree of accuracy of the Doppler flow measurement technique in the stiff tube is in agreement with earlier reports by Angelsen and Brubakk

(1976) and Gill (1979b) who used stiff tubing and a time constant flow to compare Doppler volume flow with true volume flow as measured by the amount of fluid collected in a reservoir. In our study, the cut-off level of the high pass filter was 150 Hz, so that the 29 ml min^{-1} Doppler flow overestimate in the stiff tube can be explained by the effect of filtering the low flow velocity components as described by Gill (1979b). The experiment in the elastic tube shows a high degree of accuracy between Doppler flow based on Doppler flow velocity measurements and pulsatile tube diameter as established by TD recorder, and electromagnetically measured flow (Fig. 6). The 28 ml min^{-1} Doppler flow underestimate in the elastic tube may be explained by the strong low frequency components originating from the pulsatile tubal wall which could not be completely eliminated by the high pass filters.

The axial resolution of present ultrasound equipment is until now insufficient to measure distances less than 1 mm. As a result, measurements of the tube diameter always include the outer wall thickness. This explains the Doppler overestimate of 16% when the diameter was calculated from the outer-to-inner wall distance, for which we could correct in our *in vitro* study. However, this observation is not applicable to the *in vivo* situation because with present equipment the vessel wall thickness cannot be measured and because it is not known whether the echo reflecting

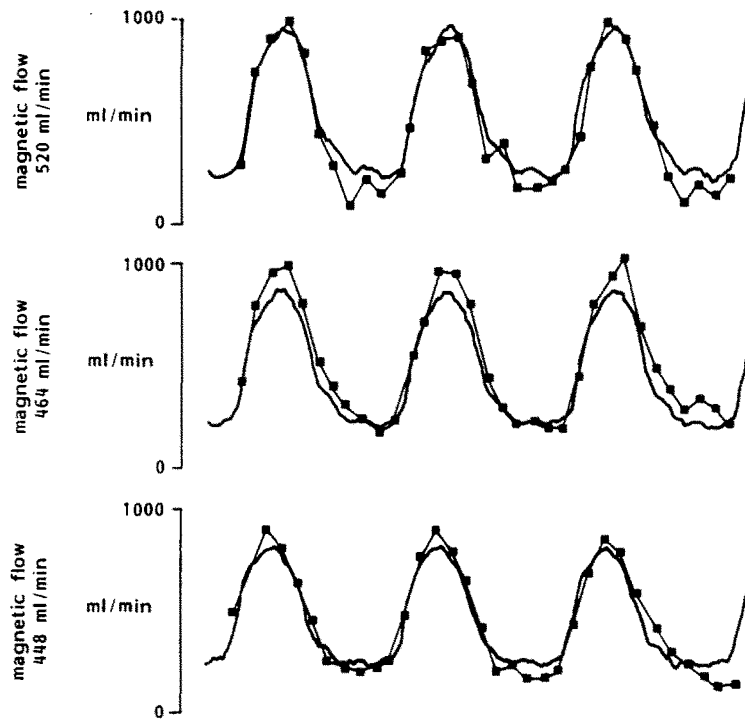


Fig. 7. Doppler volume flow values (dots) superimposed on magnetic volume flow profiles in the fetal aorta at different magnetic flow values.

boundary of the fetal aortic wall is determined by the muscular layer or by the surrounding connective tissue. We were unable to find any literature on the wall thickness of the fetal aorta. Therefore, in a pilot

study we measured the wall thickness of the fetal aorta of a normal weight 38 week old stillborn fixed at a filling pressure of 50 mmHg. In six samples we measured a mean vessel inner diameter of 6.3 (±0.7

Table 1. Percentage flow under and overestimation for the maximum (D_{max}) and minimum vessel diameter (D_{min}), the mean of the maximum and minimum diameter ($(D_{max} + D_{min})/2$) and the time-averaged vessel diameter (D_{TA}), relative to the effective diameter (D_{eff})

Patient	D_{eff} (mm)	D_{max} (mm)	Flow overest	D_{min} (mm)	Flow underest	$\frac{D_{max} + D_{min}}{2}$		D_{TA}	Flow underest
1	7.2	7.5	9%	6.5	18%	7	5%	7.0	5%
2	5.9	6.2	10%	5.3	19%	5.8	3%	5.7	7%
3	7.7	8.2	13%	7.1	15%	7.6	3%	7.5	5%
4	6.4	6.6	6%	5.7	21%	6.2	6%	6.2	6%
5	6.7	7.0	9%	6.0	20%	6.5	6%	6.5	6%
6	6.5	6.8	9%	5.8	20%	6.3	6%	6.3	6%
7	5.7	6.0	11%	5.1	20%	5.6	3%	5.6	3%
8	6.9	7.2	9%	6.2	19%	6.7	6%	6.7	6%
9	5.7	5.9	7%	5.2	17%	5.6	3%	5.6	3%
10	6.4	6.7	10%	5.7	21%	6.2	6%	6.2	6%
11	7.5	7.8	8%	6.7	20%	7.3	5%	7.2	8%
12	6.0	6.2	7%	5.4	19%	5.8	7%	5.8	7%
13	7.7	8.1	11%	7.0	17%	7.5	5%	7.5	5%
14	6.8	7.2	12%	6.0	22%	6.6	6%	6.6	6%
15	5.2	5.4	8%	4.8	15%	5.1	4%	5.1	4%
16	6.6	6.8	6%	6.1	15%	6.5	3%	6.5	3%
Mean	6.6	6.9	9.1%	5.9	18.6%	6.4	4.8%	6.4	5.4%
SD	0.7	0.8	2.1	0.7	2.3	0.7	1.4	0.7	1.5

S.D.) mm and a mean ratio between wall thickness and inner diameter of 4% (± 0.5 S.D.) when only considering the muscular layer ($0.25 \text{ mm} \pm 0.03$ S.D.) and of 5.6% (± 1.9 S.D.) when the surrounding connective tissue ($0.36 \text{ mm} \pm 0.11$ S.D.) was taken into account. Using these data we calculated an overestimate of 8.0 and 11.5%, respectively, in absolute blood flow in the fetal descending aorta. Comparison between Doppler flow and magnetic flow in the acute sheep experiment shows an overestimation in two recordings (5 and 17%) and an underestimation (-7.5%) in one recording. These results are comparable to those obtained by Eik-Nes *et al.* (1981) at flow levels between 400 and 500 ml min^{-1} in the pig aorta. It seems likely that at lower flow rates the accuracy will be even lower.

Although we studied only a comparatively narrow flow range in one fetal lamb, these data indicate that even with continuous monitoring of the pulsatile aortic wall movements by means of a TD recorder, the accuracy of Doppler flow measurements is low so that *in vivo* measurements should be interpreted with utmost care.

In the clinical part of the study (Table 1), the influence of the pulsatile aortic diameter on volume flow calculations was evaluated. We considered the effective diameter as the correct diameter, since it completely compensates for the influence of vessel pulsations on volume flow. From this it follows that, if one calculates volume flow from the maximum vessel diameter, this will lead to a volume flow overestimation of 9.1% (± 2.1 S.D.) and from the minimum vessel diameter to an underestimation of -18.6% (± 2.3 S.D.). This implies that if a diameter measurement is carried out from one frozen *B*-mode image, arbitrarily chosen within the cardiac cycle, the error in the volume flow calculation will vary between 9 and -19%. If ten randomly selected *B*-mode images are taken, the calculated mean diameter will approximate the time-averaged diameter, which in our study shows a volume flow underestimation of -5.4% (± 1.5 S.D.). From *M*-mode recordings, usually the mean of the minimum and maximum diameter is taken for volume flow calculations. For this situation, a volume flow underestimation of -4.8% (± 1.4 S.D.)

similar to that of the time-averaged diameter, can be expected.

It can be concluded that continuous recording of the pulsatile vessel diameter by a TD recorder provides a more correct approach for volume flow calculations than can be expected from *B*- and *M*-mode images. With the use of the latter methods, calculated flow values are anywhere from 19% lower to 9% higher than values obtained with the use of a TD recorder.

Acknowledgement—We would like to thank Prof. H. C. S. Wallenburg and Dr. C. E. Essed for their support in this study.

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