POWER DOPPLER

PRINCIPLES AND POTENTIAL CLINICAL APPLICATIONS

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ABSTRACT


The purpose of this work was to: a) Determine whether the amount of colour in a power Doppler image is dependent on the angle between the examined vessel and the soundbeam and/or on the velocity of the flow within the vessel; b) Investigate if a dependency on flow velocity could be used for the detection of volume flow differences and c) Define clinical applications utilising the improved sensitivity to low flow of PD.

In the experimental studies (study I and II) a silicon tube in a water bath was insonated, the insonation angle and the volume flow changed and the resulting images stored, transferred to a personal computer and analysed with regard to the amount of colour present in the image.

In study III and IV the ability of power Doppler to depict low flow was used to produce a map of the perfusion in well perfused organs, lack of colour in all or part of an organ taken as a sign of decreased perfusion. 150 patients with a renal transplant (study III) and 15 patients with abdominal trauma (study IV) were examined; the detected areas of decreased perfusion were correlated to other imaging modalities, laboratory and clinical records in order to determine the underlying pathology.

In study V the power Doppler sensitivity was used to look for and describe small portosystemic shunts in 141 patients with liver cirrhoses and suspected portal hypertension.

The colour representation in a power Doppler image was found to be dependent both on the insonation angle and the flow velocity. Computer analysis of the images could detect differences in volume flow down to a change of 10 ml/min.

Of the 150 renal transplants, areas of decreased perfusion were found in 12, all of which could be given a plausible explanation (2 local infections, 4 AV fistulae, 1 kinked segmental artery and 5 with problems related to an accessory artery).

Of the 20 organs (7 livers and 13 spleens) examined in the 15 trauma patients, 5 were found to have areas without colour, corresponding to localised haematomas. Using computed tomography as gold standard, ultrasound showed neither false positive nor false negative findings.

Of the 141 patients with cirrhosis, 40 had Doppler ultrasound findings of a shunt, consistent with a portal hypertension. 7 of these 40 shunts showed a typical “ball” or “corkscrew” pattern.

Conclusion: the colour in a power Doppler image is dependent not only on reflector concentration (as it should be in theory) but also on the insonation angle and the velocity of the flow. This can be used to detect relative changes in volume flow. Clinical applications of power Doppler include mapping of organ perfusion and the detection of small vessels. These applications are based on the high sensitivity of power Doppler.

**Key words:** Ultrasoundography; Doppler studies; Power Doppler; Volume flow.

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CONTENTS

ABSTRACT .................................................................................................................. 3

ORIGINAL STUDIES ................................................................................................... 5

INTRODUCTION ........................................................................................................ 7
  Theoretical Background ......................................................................................... 7
  Clinical Background ............................................................................................ 9

AIM ............................................................................................................................ 11
  General .................................................................................................................. 11
  Specific .................................................................................................................. 11

MATERIALS AND METHODS ............................................................................... 12
  Experimental studies .......................................................................................... 12
  Clinical studies .................................................................................................... 16

RESULTS ............................................................................................................... 20
  Experimental studies ......................................................................................... 20
  Clinical Studies .................................................................................................. 24

DISCUSSION ........................................................................................................... 26
  Specific ................................................................................................................ 26

DISCUSSION ........................................................................................................... 30
  General .................................................................................................................. 30

CONCLUSION ......................................................................................................... 32

ACKNOWLEDGEMENTS ......................................................................................... 33

REFERENCES ......................................................................................................... 34

COLOUR SUPPLEMENT ......................................................................................... 39
ORIGINAL STUDIES

STUDY I

STUDY II

STUDY III

STUDY IV

STUDY V
INTRODUCTION

THEORETICAL BACKGROUND

The Doppler effect is a change in frequency or wavelength due to motion of the source, receiver or a reflector (as is the case in ultrasound). It was first described by Christian Andreas Doppler in the middle of the 19th century but has been used by e.g. bats and dolphins to detect the motion of prey long before it was discovered and used by man. Many applications have been developed involving sound but also light and radar. However, this text is mainly about ultrasound, defined as sound with a frequency above 20 kHz (in diagnostic ultrasound usually 1 – 20 MHz). When the ultrasound is reflected or scattered by a stationary target the frequency remains unchanged but if the reflector is moving towards the source of the ultrasound the frequency will be shifted upwards and vice versa. This frequency shift can be detected by the ultrasound machine and is proportional to the velocity of the reflector. However, only the part of the flow vector that is directed towards the sound source (in our case the transducer) can be detected. Thus, if we study flow in a vessel that is not going straight to or from the transducer we need to know both the Doppler shift and the angle between the vessel and sound beam in order to determine the true velocity of the flow (fig 1). It then follows that flow cannot, at least in theory, be detected in a vessel that runs at an angle of 90 degrees to the sound beam (1).

![Diagram of Doppler effect](image)

**Fig 1a.** Every velocity can be described by a vector that has the direction of the motion and a length proportional to the speed. Then in turn, every velocity vector can be broken down into two vectors at right angles to each other. It is only the size of the vector directed at the transducer that can be detected and determined by the ultrasound machine.

![Diagram of Doppler effect](image)

**Fig 1b.** If we know the size of the vector directed towards the transducer and the angle between the measured vector and the original vector (in our case the vessel) the actual velocity of the flow within the vessel can be calculated.
The simplest forms of ultrasonic Doppler instruments use continuous waves. Echoes from tissue come back with the transmitted frequency but echoes from moving targets like the components of blood have had a frequency shift. However, as sound is transmitted and received continuously there is no way of determining where along the sound beam the moving targets are, i.e. at what depth the vessel is situated. In order to overcome this problem pulsed techniques are mainly used in our ultrasound machines. If a short pulse is transmitted and the round-trip transit time from transmit to receive can be measured, the depth of the moving target can be calculated assuming that the velocity of sound propagation in tissue is constant.

Such measurements in one small position can show the flow dynamics at this site as a continuous graph, a spectral Doppler tracing. If measurements are made at many positions all over the ultrasound image it is, of course, impossible to show the flow representation as a spectral Doppler tracing for every position. Instead, colour is used in the ultrasound image to indicate the presence of flow, most often red colour if the flow is towards and blue if the flow is away from the transducer. The hue of the colour can also be changed to indicate slow or fast flow and this is how conventional colour Doppler (CDV) works.

For abdominal work, which comprises many of the important radiological applications of ultrasound, CDV has a couple of limitations. The Doppler basics described above apply also to the colour presentation of flow and thus it is always difficult to get good and adequate colour in a vessel that runs at or close to a 90 degrees angle to the sound beam. Also, CDV is not very sensitive to low flow, especially when vessels deep in the abdomen are investigated. Combined, these limitations mean that a good colour fill-in of all portions of a curving vessel deep in the abdomen cannot be reliably counted on.

![Doppler Spectrum](image_url)

**Fig 2.** The Doppler spectrum is shifted up or down along the frequency shift axis depending on the size and direction of the flow. In a CDV presentation the position of the Doppler spectrum relative to "origo" will determine the colour in the image. When on the negative side the colour representation will be shades of blue and on the positive side shades of red. When there is no frequency shift, i.e. the Doppler spectrum around "origo", no colour will be produced. In a PD presentation on the other hand, the colour will in theory be the same in all three situations, as the size of the Doppler spectrum remains unchanged.
Power Doppler (PD) was introduced in the mid 1990’s as an alternative to CDV. Whereas on CDV the colour in the image is dependent on the shift of the Doppler spectrum along the frequency shift axis as described above (Doppler shift); PD is dependent on the number of reflectors within the sampling volume. As this parameter is unchanged regardless of the size and direction of the Doppler shift, the colour on PD should be angle independent (fig 2). PD gives no information on flow direction and, in theory, no information on flow velocity. On the other hand it offers an increased sensitivity to low flow velocities, the sensitivity being approximately 3 - 6 times that of CDV (1, 2).

**CLINICAL BACKGROUND.**

When PD was installed on the ultrasound machines for clinical work several reports soon described a so called “parenchymal blush” in well perfused organs like the kidneys and that the improved sensitivity to low flow allowed the depiction of tiny vessels both in normal and pathological tissues (3). PD could therefore be ideally suited both for a search for single tiny vessels, e.g. to differentiate viable tissue from other echogenic materials (5 6), detect hyperaemia (7, 8, 9, 10) and the assessment of perfusion in many organs (11, 12, 13, 14, 15). It was also noted on routine scanning that the amount of

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**Fig 3.**
The so-called “parenchymal blush” in a kidney. As the vessels running from the hilum to both the upper and lower poles are at a greater Doppler angle than the vessels to the central portions the poles may seem to have less perfusion. This is an artefact caused by the relative angle dependency of PD but without this knowledge serious misdiagnoses can be made. (Colour print available in the supplement, page 39).
colour in PD images varied between systole and diastole and that it seemed to be less pronounced when scanning at a high Doppler insonation angle (i.e. the angle between the vessel and the sound beam). This suggested that PD is not as independent of angle and velocity as it should be in theory. If this was the case and attempts were made at using PD to assess presence or absence of perfusion in an organ or a small vessel serious mistakes could be made if the limitations were not known and described (fig 3). On the other hand, if the colour in a PD image really did vary with changes in flow velocity and, especially, volume and this effect could be quantified it may lead to possible new applications of PD. The experimental studies were, therefore, designed to assess the true performance characteristics of PD. When this had been done so that the presence or absence of colour in a vessel or an organ (or part of an organ) could be reliably assessed taking the limitations and the characteristics of PD into account, the clinical studies were designed to determine if application of these characteristics and especially the advantages of PD over CDV could be used to extract more clinical information from the ultrasound image.
AIM

GENERAL

To examine the properties of power Doppler and describe clinical applications that utilise the advantages power Doppler has over conventional colour Doppler.

SPECIFIC

I - II.
To determine the angle dependency, if any, of the PD colour representation on the screen. To see if the clinically noted differences in the amount of colour in a PD image between systole and diastole could be related to the variations in flow volume and, if so, whether the colour differences could be quantified by a computer analysis and used to estimate blood flow volume changes.

III - IV.
To define clinical applications utilising the superior sensitivity to low flow of PD by delineating non- or hypoperfused areas in organs and relating them to pathology taking the angle dependency described in paper I into account so as not to describe technical limitations as pathological findings.

V.
To describe a typical pattern of pathology that can only reliably be seen on PD due to its sensitivity to low flow and relative insensitivity to flow angle changes compared to CDV.
MATERIALS AND METHODS

EXPERIMENTAL STUDIES

Study I

A silicon tube, selected because of its acoustic properties, was insonated with a 5 MHz linear transducer (Acuson 128XP, Mountain View, California) in PD mode through a water bath (fig 4). A blood-simulating fluid, a suspension of plastic spheres 10 mikrom, in diameter in saline (ElMeKon, Uppsala, Sweden) was pumped through the tube. The pump was modified from a radiographic film developer and gave a pulsatile flow that was calibrated at 200 and 300 ml/min by diverting the flow into a graduated glass both before and after each part of the experiment. All machine setting were kept constant during the experiments. Specifically the background was kept almost black (low mix) and a post-processing was chosen that varied the PD colour from dark to bright red.

In the actual data gathering a 2-cm long section of the tube was used, the centre of which was kept at a fixed depth with the ultrasound beam focused deeper. During the experiments the angle of insonation was altered by tilting the transducer from 90 down to 30 degrees relative to the silicon tube (the angle measures both by an external protractor and the Doppler angle device of the machine). This was first done at a flow rate of 200 ml/min and repeated at 300 ml/min. At each individual setting of flow

Fig 4.
Drawing of the experimental set-up in study I: 1, pump; 2, open vessel to air the fluid and avoid gasbubbles in the suspension; 3, water bath; 4, transducer; 5, silicone tube; 6, metal pipe to avoid sharp bends or kinks in order to maintain laminar flow; 7, connecting tubes.
and insonation angle the image was frozen, the cine loop used to find the frame with maximum colour and the image stored. 6 images were stored for each setting. Each image was then transferred to a personal computer and the amount of colour within the marked segment of the tube analysed (Adobe Photoshop, Adobe Systems, Mountain View, California). The PD post-processing (i.e. the colour scheme of the image) was chosen so that different shades of red were mainly used. This enabled us to analyse only the red video channel, which made the computations easier.

The computer produced histograms with 255 brightness levels of the red video channel. Analysing the colour in a section of background outside the tube (no PD colour) showed that the red video channel content from the background consisted mainly of darker shades up to a level of 155 whereas the PD colour mainly represented levels above 155. Thus only the levels from 156 to 255 were used for the analyses. As a brighter red colour in the chosen PD colour scheme represents more moving reflectors and to take this into account and trying to make the analysis more sensitive than a mere pixel count an integral was used. The brightness levels from 156 – 255 were divided into 10 groups (156 – 165, 166 – 175, etc), the number of pixels in each group counted and multiplied by a factor from 1 – 10 (156 – 165 x 1, 166 – 175 x 2, ….. 246 – 255 x 10). The resulting figures were added to create an arbitrary numerical value of the colour intensity in the vessel taken as a relative measurement of the flow volume. The average of the 6 images at each setting was used and the result from the different settings compared using Student’s t-test. P-values of less than 0.05 were considered statistically significant.

Study II
The principle of the set-up was the same as in study I but, as this study was designed to see if small changes in volume flow could be detected, another pump was used (fig 5). The new pump (UHDC flow Systems, London, Ontario, Canada) gave a constant flow with very small variations (a maximum of 2 %

![Diagram](image)
error in our range of flow volumes). The blood-mimicking fluid for this pump was a commercially available synthetic metal cutting fluid (Syn-Cut H.D., Acra Tech, Quebec, Canada) with acoustic properties close to those of blood. The actual reflectors, nylon copolymer particles (Orgasol 3501, Elf Atochem, Oakville, Ontario, Canada) with an average diameter in the range of blood cells, were mixed with the fluid. Again an Acuson 128XP ultrasound machine was used with a 5 MHz linear transducer. Basic machine settings were decided upon for the first experiment and then one parameter of the settings at a time was changed for the subsequent series (table 1). For each machine setting measurements were made from a low flow of 20 ml/min, then increased in steps of 10 – 50 ml/min, until most of the pixels in the marked segment of the silicon tube had a bright red colour (upper regions of the colour scale of our chosen post-processing) and each increase in flow volume resulted in only a minor change in colour (fig 6). Again, as in study I, 6 images were stored and analysed. For this study (study II) another analysis software program was used (Matrox Inspector, Matrox Graphics, Quebec, Canada) as this software made the computations easier. The mathematics of the actual calculation were, however, the same as in study I. Thus we produced a numerical value as a relative measurement of the flow volume at each machine setting and flow volume. Student’s t-test was used for the comparisons between these numerical values. P values of less than 0.05 were considered significant.

### Table 1:

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<table>
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<tr>
<td>A</td>
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</tr>
<tr>
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</tr>
<tr>
<td>E</td>
<td>Filter 4, scale +1, level 70, angle 60 degrees</td>
</tr>
<tr>
<td>F</td>
<td>Filter 1, scale 0, level 70, angle 90 degrees</td>
</tr>
</tbody>
</table>

Table 1. Basic setting (A) and the subsequent changes (underlined) made for the following experiments.
Fig 6.
Colour representation of the flow in the silicon tube at 60 degrees and a flow rate of 200 ml/min (a) and 300 ml/min (b). (Colour print available in the supplement, page 40).
CLINICAL STUDIES

Study III

The perfusion pattern, sometimes called a parenchymal blush, that PD produces in organs with a high blood flow, such as the kidneys (native or transplanted), the spleen and to some extent the liver, was used to detect areas without colour. An area without colour, provided it was not caused by the limitations of PD explored in the experimental studies, was regarded as a part of that organ with a decreased perfusion (fig 7). In this study 398 ultrasound examinations were performed on 150 patients with a transplanted kidney. They were examined either routinely within 24 hours of the transplantation procedure, when there were clinical or laboratory signs of transplant pathology or as routine check-ups in normally functioning grafts. An Acuson XP10 ultrasound machine with a 5 MHz curved array transducer was used and, depending on the findings and patient build, additional scans with either a 3.5 MHz curved array or a 5 MHz linear array transducer were performed. If the PD image produced a uniform parenchymal blush the scan was considered normal. When an area with less colour than the surrounding parenchyma was detected it was categorised as either having a decreased perfusion (colour existing but less than the rest of the kidney) or no detectable perfusion (no colour identified). A careful examination

Fig 7.
Renal transplant with an accessory artery severed at the time of transplantation. One segment of the kidney is completely without detectable perfusion while the rest of the kidney show normal PD colour.
including B-mode, spectral Doppler tracings and CDV as well as PD was performed in order to detect a reason for the PD findings. The patients clinical and laboratory records were also reviewed and correlated to possible pathology that could cause the finding.

Study IV

During the study period, 22 patients with blunt trauma to the upper abdomen were examined. Seven of these had obvious pathological findings already on B-mode imaging, such as free abdominal fluid or clear echogenic changes within the liver or spleen parenchyma. Patients with such findings were excluded from the study in order to avoid bias. The remaining 15 patients, ranging in age from 4 – 43 years (medial 11 years) were prospectively included in the study. If the trauma was bilateral or to the centre of the abdomen both the liver and spleen were examined but when the trauma was unilateral only the corresponding organ (liver or spleen) was included. All in all, 20 organs were studied, 7 livers and 13 spleens in 15 patients.

The ultrasound examinations were performed with the machine (Acuson XP10 or Acuson Sequoia 512, Siemens-Acuson, Mountain View, California, with 3.5 or 5 MHz curved array transducers) in PD mode, settings adjusted for maximum sensitivity to low flow but without blooming artefacts appearing in the image. If these settings did not produce a satisfactory parenchymal blush, mainly due to the attenuation of the ultrasound beam in a large patient, an ultrasound contrast agent was used to enhance the PD signal (Leovist, Schering, Berlin, Germany). This ultrasound contrast agent enhances the echoes from the moving blood but is a so called blood pool agent which means that it cannot produce echoes in an area of an organ that is not perfused in itself. The scans of 4 spleens and 2 livers in 5 patients were thus contrast enhanced, the agent given as a continuous intravenous infusion of 1.5 ml/min at a concentration of 300 mg/ml (20). 4 patients were given 7 ml and the 5th patient 14 ml as an extended scanning time was needed in this patient.

Each patient also had a contrast enhanced computed tomography (CT) examination performed with a standard trauma protocol using a Philips Tomoscan SR 7000. A radiographic contrast agent (Omnipaque, Nycomed, Oslo, Norway) was injected as a bolus of 300 to 400 mg iodine / kg body weight. After a 60-second delay, a spiral scan was performed and consecutive slices obtained.

The PD criterion for pathology was lack of colour in all or part of an organ defined as more than 1 cm between detectable vessels. Similarly, the CT criterion for injury was lack of contrast enhancement.

All CT and ultrasound examinations were performed within 12 hours of each other and within 24 hours of the trauma. The CT results were used as gold standard. The results of the two modalities were not compared until both examinations had been completed; thus the examiners were blinded to the result of the other modality at the time of each examination.

Study V

Patients with biopsy verified liver cirrhosis that had been examined in order to find porta-caval shunts as a sign of portal hypertension were retrospectively included in the study. During the study period 141 patients were examined. The following were considered as Doppler ultrasound signs of portal hypertension: 1) reversed flow in all or part of the portal venous system (measurements were made in the splenic vein, superior mesenteric vein, main portal vein, left gastric vein and the left and right main
portal branches) and 2) the demonstration of a portosystemic shunt (typical sites around the spleen, in or around the ligamentum teres in the liver or through the parenchyma of the left lobe of the liver). B-mode signs of portal hypertension were only used to support a diagnosis based on the Doppler signs.

The patients were examined in both the supine and left/right decubitus positions using an Acuson 128XP with a 3.5 MHz curved array transducer in B-mode as well as spectral Doppler tracings, CDV and PD. Shunts from the left portal vein were noted and classified as 1) running in or adjacent to the ligamentum teres or 2) arising from the left portal branch more peripherally and following a course through the liver parenchyma. The shunts in group “2” were further divided into those running in a straight line and those with a tortuous course, at least partly, just below the liver surface (fig 8).

Fig 8. Longitudinal sections of the left lobe of the liver show the typical pattern of a ball or corkscrew of colour just below the liver surface. This pattern is easily seen on PD but less readily on CDV due to the greater angle dependency and less sensitivity to small Doppler vectors. (Colour print available in the supplement, page 41).
RESULTS

EXPERIMENTAL STUDIES

Study 1
The average numerical values for each insona-
tion angle (30, 45, 60, 75, 80, 85 and 90 degrees) are shown in table 2 and plotted in figure 9. The values have a maximum at 45 degrees and then decrease rapidly as the angle between the ultrasound beam and the silicon tube increase. The differences between the sets of values calculated for the different angles are statistically significant (p < 0.001) both at a flow rate of 200 and 300 ml/min. There was no overlap between the sets. When comparing the different flow volumes at each angle there was a statistically significant difference between the sets of values at angles of 60 degrees and above.

<table>
<thead>
<tr>
<th>Angle (degrees)</th>
<th>200 ml/min</th>
<th>300 ml/min</th>
<th>P</th>
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<tbody>
<tr>
<td>90</td>
<td>5.67 ± 0.52</td>
<td>38.3 ± 6.7</td>
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<tr>
<td>85</td>
<td>23.5 ± 0.84</td>
<td>83.0 ± 2.0</td>
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<tr>
<td>80</td>
<td>160.7 ± 4.55</td>
<td>203.7 ± 9.85</td>
<td>0.0007</td>
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<tr>
<td>75</td>
<td>245.0 ± 11.3</td>
<td>327.0 ± 28.2</td>
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<td>365.5 ± 4.0</td>
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<tr>
<td>45</td>
<td>479.0 ± 9.0</td>
<td>481.0 ± 8.3</td>
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<tr>
<td>30</td>
<td>376.0 ± 10.4</td>
<td>393.0 ± 13.8</td>
<td>0.049</td>
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</table>

Table 2.
Average colour values for each angle and flow volume.
Study II

At the basic setting (see table 1) colour is detected when the flow rate is 30 ml/min. The curve then rises as a) more pixels within the measured region show colour and b) already colour containing pixels show colour of a progressively brighter hue (i.e. their number being multiplied by a higher factor in the integral of the computer calculations). Changing the machine settings resulted in an initial colour detection at a higher flow rate and/or a curve rising less steeply (fig 10, 11, 12 and 13). In all the machine settings, however, regardless of the colour detection level or the slope of the curve, the differences between the sets of values for each flow volume was statistically significant, p < 0.001 for the vast majority of measurements and p < 0.01 for some measurements near the high end of the curves. There was very little variation between the individual values within each set (specific machine and flow setting, 6 measurements) and no overlap between the sets.

At high volume flow rates, resulting in a high calculated value within each machine setting, there was a tendency for the curves to level off, also indicated by the slightly higher p-values of the higher flow rates in our range.
Curves resulting from the different ultrasound machine settings listed in table 1.

**Table 1:**

<table>
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<th>Filter 1, scale +1, level 0, angle 60 degrees</th>
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<td>E</td>
<td>Filter 1, scale 0, level 70, angle 90 degrees</td>
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Table 1.
Basic setting (A) and the subsequent changes (underlined) made for the following experiments.
CLINICAL STUDIES

Study III

Of the performed 398 scans in 150 patients normal parenchymal blush was seen in 186 scans in 138 patients. 12 scans (8%) showed regional differences in the amount of colour produced by PD in the renal parenchyma, 7 with colour present but decreased and 5 with no detectable colour (i.e. perfusion) in a portion of the kidney. A plausible explanation for the findings was discovered in all cases.

Diagnoses in the 7 patients with decreased perfusion were: 2 with focal infections (based on clinical, laboratory and spectral Doppler findings in addition to prompt recovery in antibiotics), 4 with arteriovenous fistulae (based on spectral Doppler and angiography findings) and 1 with a kinking of one of two arteries anastomosed end-to-side in the external iliac vein (based on spectral Doppler findings and findings at the subsequent surgical intervention).

Diagnoses in the 5 patients with no detectable perfusion were problems with small accessory arteries in all patients, 4 ligated at the time of the transplantation and 1 re-anastomosed but with doubtful patency, all according to the surgical records.

These pathological conditions were not detected by B-mode ultrasound in any of the cases. Thus, the PD examination found previously undetectable pathology in 12 out of 150 patients.

Study IV

In the PD mode, the colour fill-in of the parenchyma was deemed sufficient for a confident assessment in order to both detect and exclude hypoperfused areas in all 15 patients (i.e. all 20 organs, 7 livers and 13 spleens), 5 of whom required an injection of ultrasound contrast (4 spleens and 2 livers). The CT examinations were all of good quality.

5 ruptures or haematomas were detected, 3 splenic and 2 hepatic. All 5 were seen on both PD ultrasound and CT and there was no difference in the estimated size of the lesions between the modalities. Thus, using CT as gold standard, PD ultrasound showed neither false positive nor false negative results in this study.

Study V

The findings on ultrasound (B-mode, spectral Doppler, CDV and PD) are shown in figure 14.

Of the 141 patients included in the study, 90 patients had either no signs of PH, only B-mode signs (indirect) or inconclusive studies while the remaining 51 had signs of portal hypertension (PH) based on direct Doppler criteria. In 11 of these 51 no shunt as such was detected but reversed flow was noted in all or part of the portal venous system. Thus, in 40 patients we were able to directly visualise the shunt, rising from the left portal branch (separate from the ligamentum teres) in 10 patients. 7 of these 10 shunts showed the typical “ball” or “corkscrew” pattern, the description of which was the purpose of the study (fig 8).
Doppler ultrasound examinations for detection of PH = 141

Portal hypertension = 72
No portal hypertension = 58
  Inconclusive = 11

Diagnosis of PH based on
  Doppler criteria = 51
  Diagnosis based on B-mode
    criteria = 21

Identified shunt = 40
  Only hepatofugal flow = 11

(I) In or around the
  ligamentum teres = 16

(II) From left portal branch
  through the parenchyma = 6

(III) Extrahepatic
  = 12

(I) + (II) = 4
(I) + (III) = 2

Total from the left portal branch
  through the liver parenchyma = 10

Tortuous = 7
Straight = 3

Fig 14.
Classification of the patients included in study V.
DISCUSSION

SPECIFIC

Study I

1. In theory PD should be angle independent. We have shown that this, however, is not the case in practice when the PD is installed in the ultrasound machines and used clinically. Also, as the amount of colour in the image is supposed to be influenced only by the reflector concentration (18) within the sampling volume, a change in velocity or volume flow should not, again in theory, change the colour. An explanation is therefore needed for the dependency of the PD colour on both angle and velocity/volume shown in this study.

Most likely the effect is caused by the implementation of filters in the machines (16). This is done so that the low Doppler shift arising from normal tissue moving with e.g. respiration will not be depicted as flow. However, as most flow in normal vessels is laminar (assumed so also in this study) there is slow flow along the vessel walls. Furthermore, with an increased Doppler angle the portion of the flow vector that is directed towards the transducer will decrease regardless of the actual flow velocity. Both these factors, i.e. slow flow along the vessel walls and an increased Doppler angle will shift the Doppler spectrum towards “origo” putting a larger and larger portion of it below the cut-off level of the filter (fig 15).

2. If the calculations in this study had been based on a count of colour containing pixels only, the effect would have been based on how many sampling volumes contained one or more reflectors with a speed above the filter level and not how many reflectors there were above this level. When the flow velocity increases more reflectors within each sampling volume will move with a speed above the filter cut-off level producing a brighter shade of colour in these image pixels. To compensate for this we introduced the multiplication of the pixel count in each histogram level group by a factor 1 – 10. The ideal would have been to get a gradual increase in this multiplication rather than dividing the histogram into ten groups. As the initial calculations were made by hand this was, however, impossible.

3. The study is not blind, as the examiner was aware of the pump and angle settings. However, all computations were made after the conclusion of the entire data gathering and as the calculations are rather complex and were mainly made by the computer, bias should be minimal.

4. With the chosen post processing most of the information to be analysed was in the red video channel making the calculations easier as the software can only analyse one video channel at a time.

5. The flow in the silicon tube was considered laminar due to the length of the connecting tubes and the absence of sharp bends in the set-up. The assumption is supported by the fact that there was little or no colour in the image at a Doppler angle of 90 degrees. Had there been turbulent flow the ultrasound machine would most likely have picked up the lateral movement of swirling reflectors.

6. As mentioned above (results, study II) the amount of colour can only increase until all pixels in the measured area contain colour of the maximum brightness. This is the probable reason for our inability to detect differences between 200 and 300 ml/min at 45 degrees.

7. Even if all pixels contain maximum colour, the values for 30 degrees should have been as high as those for 45 degrees. That they were not is
Fig 15.
When the detectable flow vector decreases (decreased flow velocity or increased Doppler angle) the Doppler spectrum is shifted towards “origo” and will be partly cut off by the filter.
thought to be caused by an artefact that appeared as dropout of colour in small areas at 30 degrees. This artefact has been, unfortunately unofficially, described and is, according to the manufacturer, caused by oversaturation of the image-forming process in the machine.

8. At 90 degrees there is very little colour in the image regardless of flow volume. This and the differences in colour shown (at angles of 60 degrees and above) when the flow rate is changed would indicate that, at least in this setting, the colour presentation of PD on the screen, here represented by the computer calculated values, is influenced both by the Doppler angle and the flow volume (17).

Study II

1. Even if the long term goal of volume flow analyses would be to determine flow in capillary beds in the parenchyma of an organ PD colour is well suited also for analysis in a single vessel as the ability to use only one video channel makes the calculations much easier. It also provided, for this experiment, an easily reproducible model of high accuracy, which would have been difficult to accomplish should we have used a more capillary bed-like set-up.

2. The variations in slope and basic detection level seen between the measurements at different machine setting are probably mainly due to changes in the cut-off level of the filters in the machine. The amount of colour assigned to a certain Doppler spectrum energy is also influenced by the scale and level settings (amplification) whilst an increase in the Doppler angle will simulate a decrease in velocity as the flow vector directed towards the transducer will decrease.

3. As in study I the examiner was not blinded to the machine and pump settings but the same principles as mentioned above are thought to apply.

4. The curves have a tendency to level off at high flow volumes when all pixels have acquired a hue approaching level 255 in the histogram. This would be the expected result as no more colour can be produced when all pixels within the measured region already contain colour of bright shades. The machine will deal with a further increase in flow by adding a signal in the green video channel. We have not analysed this information as it would have been difficult to accurately assign a value to the green colour. If, on the other hand, the information on which the colour is produced could be extracted from the machine it is probable that the steeper parts of the curves could be extended.

5. The size of the vessel (functionally) is determined by the number of pixels with colour and their colour hue is dependent on the velocity. As the computer analysis will result in an arbitrary number as an indication of the volume flow rather than an absolute figure in ml/min we are at the moment limited to measuring volume flow changes, ideally in a specified position, serially in the same patient. Doing this would also overcome the problem with attenuation of the sound beam on its way to the region to be measured (18, 19). When measuring in a single large vessel the angle of insonation must be taken into account and standardised if comparison is to be made with previous examinations.

It must therefore be remembered that what the study shows is that changes in flow volume can be detected and quantified. Volumes cannot be given as ml/min.

6. Different machine settings can be used to adapt the set-up so that the highest sensitivity corresponds to the expected blood flow in the vessel to be examined.

Study III

1. Several of the causes for regional differences detected by PD in this study also have criteria on e.g. spectral Doppler but finding the affected
area by PD makes it unnecessary to look through the entire renal parenchyma in search of possible spectral Doppler signs (measured only at one site at a time).

2. A previous study used CDV to search for underperfused regions in transplanted kidneys (11). This can of course be done but the level of accuracy for detection of small affected areas as well as the ability to examine deeply situated kidneys is potentially improved by using PD.

3. When the feeding artery to a portion of the renal parenchyma is stenosed or occluded it is obvious that the perfusion will be decreased. The reason for the decrease in the cases of focal infections and AV fistulae is less apparent but is believed to have been caused by a steal phenomenon (AV fistulae) and a localised oedema (infections).

4. The PD examination found otherwise easily overlooked pathology in 12 out of 150 patients and, at least in 7 of these 12, pathology that could be crucial to graft survival.

Study IV

1. During the first days after a trauma a haematoma in the parenchyma of the spleen and liver may be completely isoechoic on B-mode ultrasound compared to the surrounding tissue. This has limited the role of ultrasound in the acute management of trauma to the detection of free abdominal fluid in major traumas.

2. Contrast enhanced CT is a well-tested and reliable method for the detection of organ laceration. On the other hand there are many patients who cannot be examined with CT, e.g. patients with previous allergic reactions to contrast agents or patients who cannot lie still. If a reliable method utilising ultrasound could be developed it would offer an alternative for these patients and possibly for patients, especially children, with a localised trauma where the radiation of an entire trauma-protocol CT may be unnecessary.

3. As can be derived from the ages of our study population many of these patients fall into the above-mentioned category. There is an overrepresentation of minor trauma in children reflecting both the geographical situation of the hospital (no major motorways with through traffic) and, possibly to a greater extent, the fact that major trauma often produce pathology of a degree that makes immediate surgery necessary or findings that would be obvious also on B-mode ultrasound, the latter being an exclusion criterion.

Study V

1. Early stages of cirrhosis may go undetected by B-mode ultrasound and this place added importance on knowledge and detection of Doppler signs of flow pattern changes (21, 22, 23).

2. Shunts from the left portal branch may exist in up to 34 % of patients with an established portal hypertension (24). Indeed, in 7 out of 50 patients in this study such shunts were the only detectable sign making it even more important to detect them.

3. Flow in a shunt can be detected by CDV as well as PD. Looking through the entire left lobe of the liver with the machine set for a high sensitivity and tissue motion artefacts from the adjacent heart can be both difficult and time consuming. Recognising the typical pattern here described may help to find shunts otherwise missed and this particular pattern is not as easily seen with CDV as with PD due to the greater sensitivity to low flow and flow at high Doppler angles of PD.

4. The diagnosis of portal hypertension in this study is based solely on the ultrasound findings with no corroborative investigations. However, most of the direct Doppler signs are clear-cut and we consider the demonstration of a shunt in a patient with a biopsy verified cirrhosis as highly suggestive in itself. 20 of our patients have since the study developed clinical signs of portal hypertension.
DISCUSSION

GENERAL

There are potentially numerous applications for a modality that can detect blood flow in small vessels with a high sensitivity, with or without contrast enhancement (24, 25, 26, 27, 28, 29). PD has a sensitivity that is reported to be up to six times that of CDV enabling the examiner to detect, follow and determine the flow in small vessels. The vessel anatomy is easier to assess with a more complete colour fill-in of all vessels in the region. Perfusion applications range from the simple proving that perfusion exists, via perfusion estimation (30, 31, 32, 33), up to complex quantifications of blood flow, subjective (34, 35, 36, 37) or objective (38, 39, 40, 41, 42, 43, 44, 45). Finding a single vessel and by doing so differentiating viable tissue from other echogenic materials like gall bladder sludge, necrotic material in an abscess or a haemorrhage will give valuable information that may lead to earlier diagnosis of tumour tissue and the avoidance of attempted drainage procedures directed at solid areas. When the high sensitivity of PD is utilised to produce a parenchymal blush, small areas without perfusion can be detected whether they represent tissue without perfusion (e.g. emboli) or the replacement of normal tissue by isoechogenic material (haematomas). In well perfused organs like liver, spleen and, especially, the kidneys this can be used in almost the same way CT uses radiographic contrast, i.e. less contrast equals less perfusion. When the attenuation of the ultrasound beam as it passes through tissue is too great for an acceptable degree of parenchymal blush to appear in the studied organ, an ultrasound contrast agent may be used to enhance the Doppler signal. Recently, advances in technology using ultrasound contrast have made it possible to selectively depict the contrast particles in a B-mode image. This may further enhance the role of ultrasound in the detection of hypo- or non-perfused areas of an organ. PD will, however, always have the advantage over contrast enhanced ultrasound that it is non invasive and readily accessible, i.e. if the clinical question can be answered by simply pushing the PD button on the machine there will be no need to extend the examination to the use of contrast.

Quantification of blood flow, be it in a single vessel or a capillary bed would have many obvious advantages and clinical applications. Attempts have been made with time domain processing (46) or calculations from flow velocity and the cross sectional area of the vessel (47). The interobserver variation of such calculations is, however, considerable (48). Analysis of the PD colour may offer another path to explore keeping in mind that such analysis, be it subjective or objective, never can give a result in ml/min. It is, however, very sensitive to small changes in blood flow and there are many instances where the relative change of flow is much more important than the absolute figures.

Subjective assessment of the PD representation of flow has been used clinically to determine hypaemia in inflammatory tissue (e.g. joints and tendons, inflammatory bowel diseases), to assess tumour blood flow or to differentiate highly vascular focal lesions, e.g. in the liver.

Objective assessment is possible by either analysing the image (as we have done) or, much better, by getting access to the raw data on which the PD colour is based in the machine. It has been done clinically in order to quantify tumour blood flow or to detect a decreasing flow in a single vessel as a sign of a developing stenosis. In our in vitro experiment we could detect differences in flow volume down to a change of 10 ml/min. Changing the machine settings could shift the steepest part of the curve, i.e.
the region with the greatest sensitivity to flow changes, so that a study could be adapted to the expected flow in the vessel to be examined. Even taking into account the ultrasonographically ideal conditions when scanning in a water bath and the often not so ideal conditions in vivo the sensitivity to flow changes should be such that it is enough for most clinical applications.

When PD was first introduced it was, based on a theoretical reasoning, reported to be angle independent. It is, in clinical practice, obvious that PD is less dependent on the Doppler angle than CDV but also that it is not completely independent on neither the angle nor the actual flow in the vessel. If, for example, flow cannot be detected in a vessel the examiner must be able to make a knowledgeable assessment of whether this finding actually represents a lack of flow or simply a technical limitation. Knowledge of these dependencies is crucial if PD is to be used for any but the most simple clinical applications and indeed serious mistakes can be made if quantifications are made of images taken from different angles.

We set out to determine the PD degree of dependency on the Doppler angle (study I), the dependency on flow volume/velocity and whether this dependency could be used to our advantage (study II). This done, so that a PD image could be interpreted with knowledge of the pitfalls, we developed clinical applications utilising the improved sensitivity to flow in a capillary bed (study III and IV) and in a small single vessel (study V).
CONCLUSION

The PD image colour representation of flow is dependent on both the Doppler angle and the velocity of the flow in the vessel to be investigated. The former characteristic needs to be know and taken into account in clinical ultrasound scanning while the latter can be used to detect small changes in blood flow volume in a single vessel.

Taking these determined characteristics of PD into account the high sensitivity can be used to detect perfusion defects in otherwise well perfused organs enabling us to use ultrasound for new areas of image based diagnoses or to find and follow the path of vessels difficult to detect with CDV.
I dedicate this work to my parents, Göte and Ingrid, without whom I would not have been here, neither literally nor metaphorically; and to my family, Vicky, Chris, Alex and Freddie, without whom being here would have been pointless anyway.

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COLOUR SUPPLEMENT

Colour figures from thesis and from papers originally printed in colour.

Thesis - fig. 3.
Thesis - fig. 6. and Paper I - fig. 2.
Thesis - fig. 8. and Paper V - fig. 2.
Paper IV - fig. 1b.