

# Methods for the Evaluation of Vascular Reconstruction

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## Key Words

Arterial reconstruction · Doppler flowmetry · Electromagnetic flowmetry · Flowmetry · Intravascular ultrasound · Outflow resistance · Pressure recording · Transit time flowmetry · Ultrasound

## Abstract

Despite carefully performed surgery, vascular defects can remain undetected within an arterial reconstruction after blood flow is restored. Failure to recognize technical errors or abnormalities can lead to vessel thrombosis and embolization. For cardiac, carotid or visceral reconstructions these events can be catastrophic. Furthermore, corrections of lesions before graft thrombosis can have a significant impact on long-term patency, being particularly important for vein grafts, because most will not maintain patency with thrombectomy alone. This review describes methods for peroperative evaluation of vascular reconstructions. Since careful selection of patients for therapy as well as proper postoperative surveillance contribute to the success of vascular surgery, some methods for pre- and postoperative evaluation are also included.

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## Introduction

The need to measure blood flow intraoperatively became apparent during the development of reconstructive arterial surgery. However, many surgeons unfortunately still rely on pulse palpation as an index of flow, not realizing that a vessel can pulsate even when there is no blood flowing through it. In fact, if the vessel occludes distal to the palpation site, the pulse may even increase, though admittedly not for long (fig. 1).

The primary aim in flowmetry is to obtain information on the immediate results of the reconstruction, where a technical failure may jeopardize an otherwise successful operation. There is a clear correlation between the blood flow values obtained following injection of papaverine and the prognosis of the arterial reconstruction. Dedichen [1] found in patients with lower limb arterial reconstruction that the risk of early postoperative occlusion was significantly increased if the basal blood flow after reconstruction was less than 100 ml/min or the papaverine-induced flow was less than 200 ml/min. However, it can be questioned whether the flow measurements are sensitive enough to detect minor technical failures since normal blood flow values have been measured despite extensive defects at the anastomosis [2]. Furthermore, the levels of blood flow recorded during an operation may have little relation to normal activity. Under some type of anesthesia and conditions due to acute blood loss, blood flow

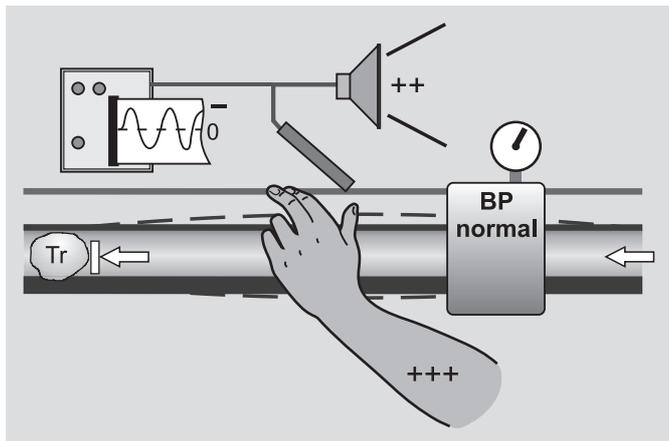
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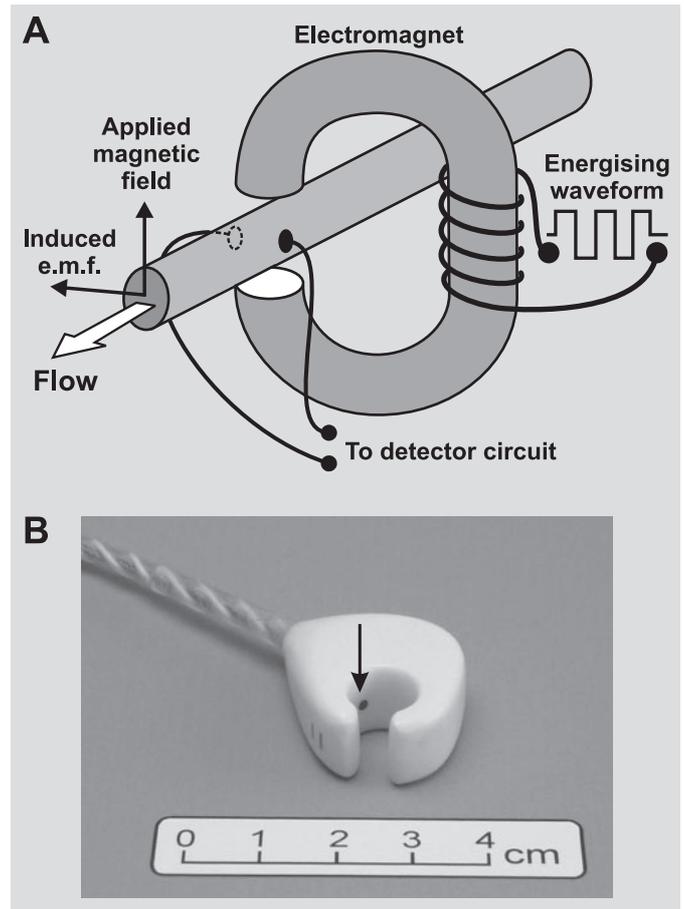
**Fig. 1.** Pulse palpation is a poor indicator of blood flow. If the artery is occluded distal to the palpation, the pulse is increased compared to an open artery. Because the artery is compliant, there is some blood flow back and forth into the occluded segment. This movement might produce a Doppler sound interpreted as normal, whereas the spectrum more correctly shows a reverberant flow with zero mean blood flow. BP = Blood pressure.

is likely to be markedly depressed, whereas hyperemia is observed in another type (epidural anesthesia).

The *basal* blood flow after bypass grafting is of limited help to precisely evaluate the arterial reconstruction, since it depends on various factors in addition to the graft itself, e.g. arteriolar tone, blood volume, body temperature and the type of anesthesia. The blood flow increase observed following administration of a vasodilating agent, e.g. papaverine or iloprost [3–5], is a significantly better indicator of the long-term prognosis, and is better suited for the detection of technical failures.

Five-year patency of femorodistal bypasses varies between 40 and 70% in many series. Some of the reclosures may be due to intimal hyperplasia, whereas others are caused by technical failures leading to secondary thrombosis formation and graft occlusion. To eliminate technical failures, intraoperative control of the reconstruction is important. Furthermore, such investigations can also be helpful in planning the procedure, and in giving an indication of the long-term prognosis of the operation.

In cardiac surgery, flowmetry was introduced at an early stage to indicate the severity of disease, and to assess patency of coronary bypass grafting and the prognosis of therapy [6–12]. In coronary surgery, flowmetry is still regarded as an essential tool for evaluating the anastomotic sites [13, 14], and guidelines for performing and interpreting flowmetry procedures have been developed [15].



**Fig. 2.** **A** The principle of an electromagnetic flowmeter. e.m.f. = Electromotive force. **B** The electromagnetic flow probe for 8-mm vessels (Nycotron electromagnetic flowmeter). The arrow points at one of the two surface electrodes.

The purpose of this presentation is to review methods for evaluation of arterial reconstructions, and to discuss which methods are going to be important in the future.

### Functional Tests

#### *Electromagnetic Flowmeter*

Electromagnetic flowmetry quantifies blood flow in a single vessel. The technique is based on the principles of electromagnetic induction, described by William Faraday 170 years ago.

If an electrolyte (such as blood) flows at right angles to a magnetic field, then an electromotive force is induced in a plane, which is mutually perpendicular to the magnetic field, and to the direction of fluid flow (fig. 2). Two electrodes situated in the appropriate plane and connected to

a suitable detector circuit can measure the induced voltage, which is proportional to the strength of the magnetic field and to the velocity of blood flow within the blood vessel. For most clinical and experimental applications, the probe is C-shaped so that it can be slipped around the vessel. The probe must be carefully chosen so that it fits tightly round the vessel to ensure a good contact between the electrodes and the vessel wall, but does not compress it. The probe is designed to produce an even magnetic field, ensuring accurate measurements. Incorporated in most electromagnetic flowmeter transducers is the electromagnet in addition to the two recording electrodes (fig. 2B). As the probes are manufactured for specific vessel diameters, a set of differently sized probes is needed. These are usually calibrated to give blood flow (ml/min). Most flowmeters utilize an alternating magnetic field to avoid polarization of the detector electrodes [16].

One important factor is zero stability [17]. When recording from peripheral vessels occluding the vessel with a clamp can check the zero reading. However, this is often impossible when flowmeters are used on the aorta or pulmonary artery.

Another source of error in making electromagnetic flow measurements is the variation in sensitivity that is observed with changing hematocrit [18, 19], influencing several commercial flowmeters. Consequently, whenever possible the instrument should be calibrated for the particular patient. The sensitivity to hematocrit is largely due to the changing electrical impedance of blood, which also changes when blood is moved from a state of rest, as the erythrocytes tend to move to the center of the vessel (axial accumulation). This supposedly leaves a plasma layer at the vessel wall.

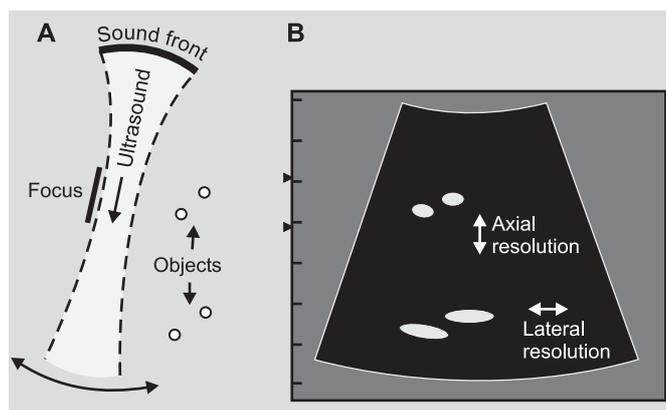
Recording of blood flow in expanded polytetrafluoroethylene (PTFE) grafts with electromagnetic flowmetry is limited by the electrical isolation of the graft material [20]. In patients with such graft material, measurements are usually performed in the native vessel proximal to the proximal anastomosis or distal to the distal anastomosis.

#### *Ultrasound Techniques*

Sound waves are pulsating pressure waves, and when above audible, e.g. above 20,000 Hz, they are termed 'ultrasound'.

The sound is characterized by:

- *Intensity*, i.e. sound pressure or pressure amplitude.
- *Frequency*, i.e. number of oscillations per second (Hz). Audible sound is between 20 and 20,000 Hz. Medical ultrasound is normally ranging between 2 and 15 MHz (= 2–15 million Hz), although intravascular ultrasound



**Fig. 3.** Spatial resolution in ultrasound consists of axial and lateral resolutions. The imaging of point targets results in a smeared-out image. Note that the lateral resolution is poorer than the axial resolution, and that it is dependent on depth, unlike axial resolution (re-drawn from Angelsen [21]).

(IVUS) equipment may use ultrasound frequencies of 30 MHz.

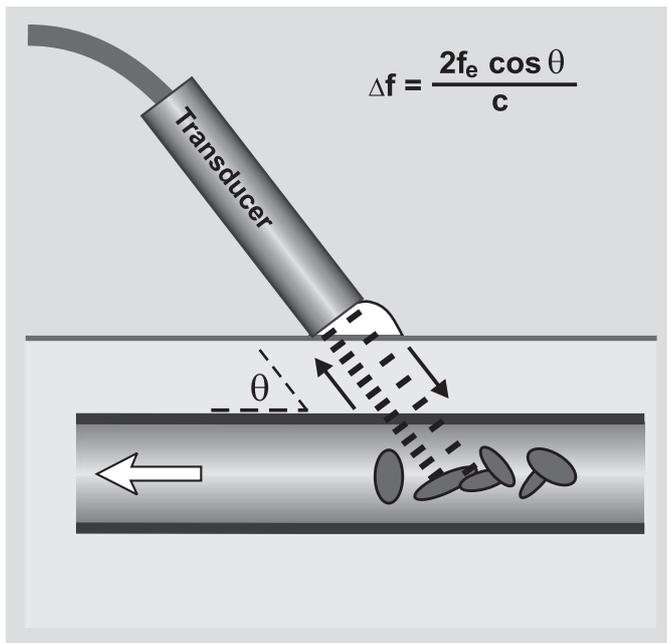
- *Wavelength ( $\lambda$ )*, i.e. distance between two adjacent maximal or minimal sound pressure levels.

An important property of the medium where the sound propagates is the specific ultrasound velocity ( $c$ ). In air this is approximately 340 m/s, in water approximately 1,500 m/s. Human tissue is similar to water and the sound velocity is approximately 1,540 m/s. The wavelength is the ratio between velocity and frequency ( $f$ ):  $\lambda = c/f$ . For medical ultrasound in human tissue  $\lambda$  is 0.15 – 0.75 mm (for audible sound  $\lambda$  is between 2 cm and 20 m), dependent on the frequency used.

Ultrasound energy is generated by the use of piezoelectric ceramics with electric wires on each side (transducer). The ceramic is compressed and expanded in parallel with an applied oscillating voltage, acting like a loudspeaker. On the other hand, if the ceramic is compressed and expanded by a sound wave, a variation in voltage is generated, and the ceramic acts as a microphone.

#### *Resolution*

Resolution is the ability to detect details. In an ultrasound system there is a clear distinction between axial (along the sound wave) and lateral resolution (90° to the sound wave). Axial resolution is usually higher than lateral resolution. It is dependent on the frequency of the ultrasound emitted (the higher the frequency, the more detailed is the image obtained), but independent of the depth of the



**Fig. 4.** Schematic drawing of a Doppler velocity detector. An ultrasound beam is transmitted by the transducer. Echoes from particles that have a movement relative to the transducer contain frequencies different from the transmitted frequency. The change in frequency, the Doppler shift ( $\Delta f$ ), is given by the formula, where  $f_e$  is the emitted frequency,  $v$  is the velocity of the scatters,  $\theta$  is the angle between the ultrasound beam and the velocity vectors in the vessel and  $c$  is the speed of sound in the tissue.

measurement. Lateral resolution is dependent on wave length and probe size, and decreases (lateral point-size increases) with increasing depth (fig. 3). Consequently, the ultrasound image is sharpest in the focus area of the transducer, and gets more blurred in larger depths [21]. Nowadays multiple focus zones overcome this problem.

#### Signal Attenuation

The signal *intensity* depends on the attenuation of the ultrasound energy as it passes through the tissue. The attenuation is dependent on *reflection*, *scattering* and *absorption*.

**Reflection.** Reflection occurs at any site with different acoustical impedance. The greater the difference in impedance, the stronger is the reflection. This occurs in the boundary layers between tissue and bone and air, respectively, for instance against bowel gas. Ultrasound gel is used to eliminate the layer of air between the transducer and the skin surface, thereby avoiding reflection of the ultrasound. However, reflection per se is not always undesirable. In fact, a reflection from boundaries within the

tissue is the very basis for generating images with anatomical information in B-mode scanners.

**Scattering.** When a reflecting object is small compared to the ultrasound wavelength, a portion of the wave energy is scattered in all directions. As scattering is proportional to the 4th power of the frequency, the attenuation increases steeply with increasing frequency.

**Absorption.** As the sound wave travels through tissue, some of the energy is converted to heat by absorption. Absorption is proportional to the 1st power of the frequency.

The selection of a specific frequency for various diagnostic procedures always involves a compromise between the optimal ultrasound penetration depth (lower frequency – higher penetration depth) and the ability to identify relevant tissue changes (higher frequency – better sharpness or resolution). For cardiology and abdominal examinations, transducers with frequencies between 2.5 and 3.5 MHz are chosen, for obstetrics 3.5–5.0 MHz, and for peripheral vessel examinations 5.0- to 15.0-MHz probes are normally preferred.

#### The Doppler Principle

When an object moves relative to a sound source transmitting a given frequency (e.g. an ultrasound transducer), the frequency of the backscattered ultrasound is different from the one emitted: the frequency is higher if the movement is towards the transducer, and lower if the movement is away from the transducer (fig. 4). This change in frequency is called Doppler shift ( $\Delta f$ ) and is described by the equation:

$$\Delta f = 2 f_e \cdot v \cdot \cos \theta / c$$

where  $f_e$  is the emitted frequency,  $v$  is the velocity of the moving objects (usually red blood cells in circulating blood),  $\theta$  is the angle between the ultrasound beam and the vascular axis, and  $c$  is the speed of sound in the tissue (approximately 1,540 m/s). When using a pencil-probe Doppler transducer,  $\theta$  is usually not known and may represent an error in recording blood velocity. In duplex scanners, however,  $\theta$  is compensated by visual angle correction, and the velocity calculation is thus more accurate.

Small, 'pocket Doppler' units are usually continuous wave (CW) Doppler velocity detectors. These include two piezoelectric elements with overlapping ultrasound fields, one that *continuously* emits ultrasound, whereas the other is *continuously* detecting sound waves (fig. 5). The overlap fields represent the area where Doppler signals may be detected, and is termed *sample volume*. CW Dopplers have no range resolution, and cannot distinguish between signals from different vessels lying along the path of the beam.

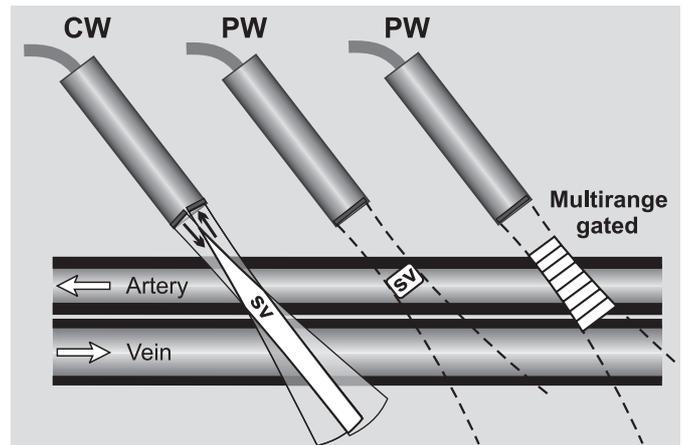
Pulsed wave (PW) Doppler systems are used to detect blood velocities at specific depth. Using only one piezoelectric element, the transducer functions alternating as transducer and receiver. The signal received is sampled at an adjustable delay after pulse transmission. This way, signals at a given tissue depth are selected (sample volume, fig. 5). The sample volume size is variable, and is determined by both the characteristics of the ultrasound beam and by the duration of the ultrasound pulse. Shorter pulse length (shorter ‘packets’) during the transmission phase results in a smaller sample volume, with a corresponding increase in the axial resolution [22].

The pulsing beam may, however, introduce a problem when measuring large blood velocities, termed frequency aliasing. To avoid aliasing, the Doppler shift must be less than half the pulse repetition frequency, PRF (often referred to as the Nyquist limit of the frequency). A new transmission pulse is not allowed until the former pulse is sampled. At larger sampling depths, the time between transmission and gating has to be longer, because of the longer time needed for the sound wave to reach the target area and to return to the transducer. In turn, this increases the time between the pulses, i.e. reduces PRF, and thereby reduces the limit where aliasing occurs. To correctly detect very high velocities at large depths, for instance in mitral stenoses, it may be necessary to switch to CW to avoid aliasing problems.

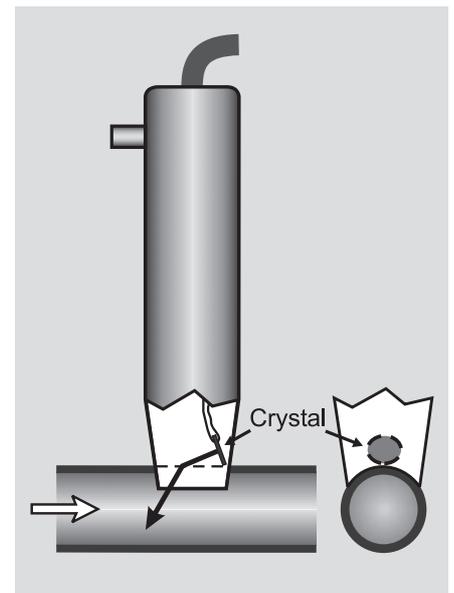
#### Doppler Flowmeter

In Doppler flowmeters, the mean blood velocity (cm/s; the ‘mean blood velocity’ refers to the temporal average of the mean velocity vector trace) is multiplied by the internal cross-sectional area (cm<sup>2</sup>) to obtain the calculated blood flow (ml/min). This calculation is made automatically by some flowmeters by a multirange-gated ultrasonic beam where the diameter is derived, or the flow is obtained semi-automatically with an on-line analogue calculation unit where internal diameter is manually inserted. In both techniques, the accuracy of the calculated volume flow is highly dependent upon the correct determination of the diameter because the error is squared at the calculation ( $Q = v \cdot \pi \cdot r^2$ ), where  $v$  is velocity and  $r$  the internal radius of the vessel. For example, if the radius of an artery of 3.0 mm is erroneously taken as 3.5 mm, the calculated blood flow is overestimated by 35%.

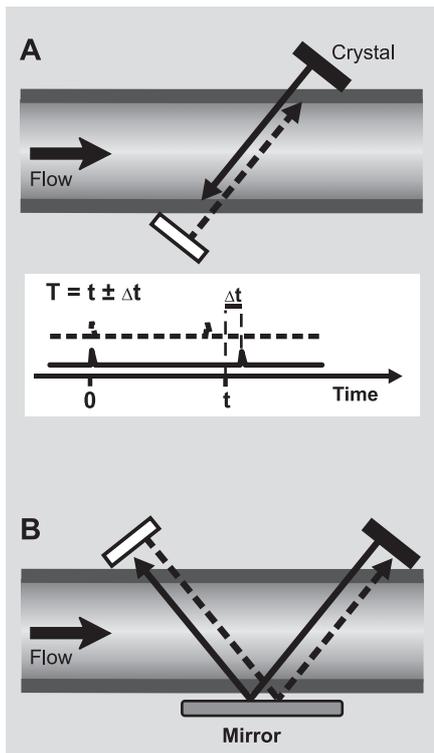
To minimize errors associated with angle insonation and vessel diameter, the transducer may be fixed to the vessel wall with a special cuff [23], or by customizing special transducers with inbuilt angle correction (fig. 6). Doppler flowmetry has the advantages over the electro-



**Fig. 5.** Schematic illustration of CW, PW and multirange-gated PW Doppler techniques. CW Dopplers contain two piezoelectric crystals: one that continuously emits ultrasound while the other continuously detects ultrasound echoes from the tissue. The overlap field is the location where Doppler signals may be detected, the ‘sample volume’ (SV). The sample volume is smaller in PW Dopplers, and the SV depth may be selected. In multirange-gated PW Dopplers the SV is sectioned, enabling identification of various Doppler spectra through the cross-section of the vessel.



**Fig. 6.** Doppler ultrasound transducer made for intraoperative use (developed by the author). The concave tip ensures good contact with the vessel wall even without a coupling medium. The transducer is placed at 90° to the vessel with the direction identifier at the top pointing upstream. The inbuilt crystal angle correction ensures fixed vessel insonation at 60°.



**Fig. 7.** The principle of transit time flowmetry (**A**). Ultrasound is alternately sent upstream (continuous line) and downstream (interrupted line). When blood is not flowing, the passage time ( $T$ ) is  $t$ . The passage time is minutely ( $\Delta t$ ) increased when sent upstream and reduced by the same amount when sent downstream. The magnitude of  $\Delta t$  is dependent on blood flow. In practice, the transducer crystals are placed on the same side and the ultrasound beam is reflected at a mirror (dual pass; **B**).

magnetic flowmeter of requiring no zeroing, and it is virtually unaffected by the vessel wall provided no air is entrapped within the wall material, for instance at newly inserted expanded PTFE and Dacron grafts. This problem might be reduced by gently squeezing the PTFE graft between two fingers for 2–5 min, enabling blood to penetrate the porous graft material [24].

#### *Transit Time Flowmeter*

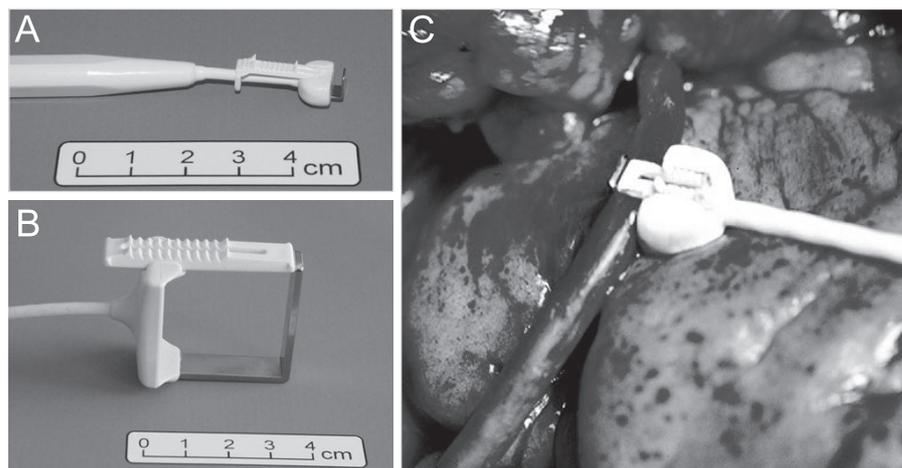
The most recently developed technique is the transit time flowmeter, also using ultrasound, but not the Doppler principle. This method is based on the fact that ultrasound traveling against the bloodstream will take a longer time than when moving downstream (fig. 7A). The theoretical basis for this technique has been known for some time, but only in the latest decade practical solutions have been available. An example is the Medi-Stim VeriQ flowmeter (fig. 8; [www.medistim.com](http://www.medistim.com)), including a computer-



**Fig. 8.** A computer-based transit time flowmeter, Medi-Stim VeriQ.

based system with trend functions and the ability to measure peripheral vascular resistance. Transit time flowmeters are very accurate and reproducible, even at minute flow rates, and possess high zero line stability [25, 26].

A practical transducer design is shown in figures 7B and 9, where the two ultrasound crystals are placed on one side of the vessel and a metal reflector on the opposite side. This induces twofold passage of ultrasound through the blood vessels. The downstream crystal transmits a pulse of ultrasound to the upstream crystal. The difference in transit time propagation depends on the blood flow. The blood flow prolongs the time for the ultrasound beam to pass through the vessel against the bloodstream, and a phase detector senses the time difference. A new sound beam is sent downstream and reflected. The blood flow decreases the transit time because the transmitting medium is also moving downstream. The process is then reversed several hundred times a second. Since the position of the crystals is fixed, the difference in transit time between the two directions can be related directly to blood



**Fig. 9.** The two piezoelectric crystals of transit time transducers are placed within the housing on one side of the vessel, with a metal reflector on the opposite side. Transducer for small-caliber vessels (**A**), e.g. coronary bypasses, and for large vessels (**B**), e.g. aorta, are shown. **C** Measurement of blood flow in a saphenous vein coronary bypass.

flow. An extremely sensitive and stable detecting device is required to measure the very small difference between the transit times, which is in the order of picoseconds ( $10^{-12}$  s), but the method has the advantage that the flow direction, as well as the volume flow, is indicated. A prerequisite for a correct estimate of blood flow is even insonation of the cross-section of the vessel. On the other hand, the difference in the transit times is only dependent on the moving particles in the vessel, thus making the measurements independent of the inner diameter.

At our hospital, the use of intraoperative flowmetry has been implemented in peripheral vascular procedures for about 35 years. In this period, all the described techniques have been employed: In the 60s and 70s, electromagnetic flowmetry was used, in the 80s ultrasound Doppler technique with custom-made peroperative transducers, and in the 90s we shifted to transit time flowmetry, which we regard as the method of choice for peroperative blood flow measurement.

#### *Examples of Flowmetry in Peripheral Vascular Disease*

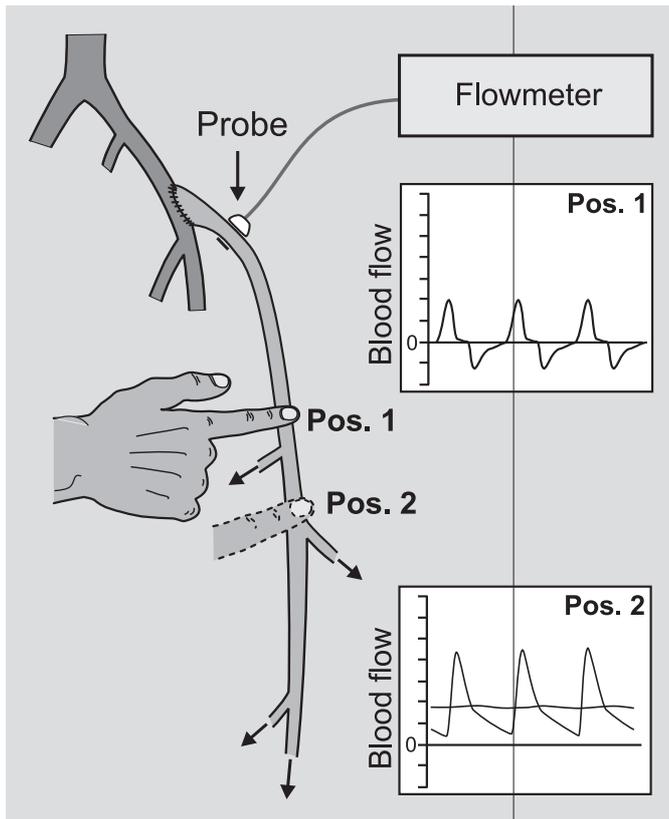
*Detection of Side Branches during 'in situ' Bypass Surgery.* During in situ bypass surgery, the side branches of the great saphenous vein must be interrupted to prevent them from becoming arteriovenous fistulae after the reconstruction. Following completion of the proximal and distal anastomoses, vessel clamps are removed and blood flow through the bypass is established. Major tributaries of the vein can be visually identified and ligated without difficulty. However, smaller branches often have to be detected by other means, mainly intraoperative flowmetry, in order to avoid major surgical dissection. It is

important to locate these residual fistulae since they may cause graft failure [27].

The detection of arteriovenous connections has a qualitative aim; exact blood flow values are less relevant. A flow probe of appropriate size is placed around the vein, just below the proximal anastomosis. When using transit time flowmetry, either sterile saline or ultrasound gel is used to maintain acoustic coupling between the probe and the measuring site. Pulsatile flow curves will immediately appear on the display (fig. 10). Manually or by means of an atraumatic clamp, the vein is occluded successively at different sites along its course, from the transducer to the distal anastomosis.

A reverberating flow profile, indicating no net blood flow (fig. 10, position 1), is found when there is no leakage flow between transducer and the compression site. If flow is detected during clamping, we can expect to find an open side branch proximal to the site of clamping (fig. 10, position 2), which can subsequently be ligated.

*Intraoperative Functional Evaluation of a Vascular Reconstruction.* As distal femorotibial bypass grafting has become more common, supplementary methods are necessary for intraoperative control. The primary aim is to obtain information on the prognosis for the immediate result of the reconstruction. Several studies have shown that peroperative flow values have prognostic values [1, 3, 5]. The risk of early postoperative occlusion is significantly increased if the basal blood flow after femoropopliteal reconstruction is less than 100 ml/min or the papaverine-induced flow (intra-arterial injection of 40 mg papaverine) is less than 200 ml/min (fig. 11). The effect of papaverine is reduced if the surgery is performed under epidural anesthesia, since basal flow is already increased.

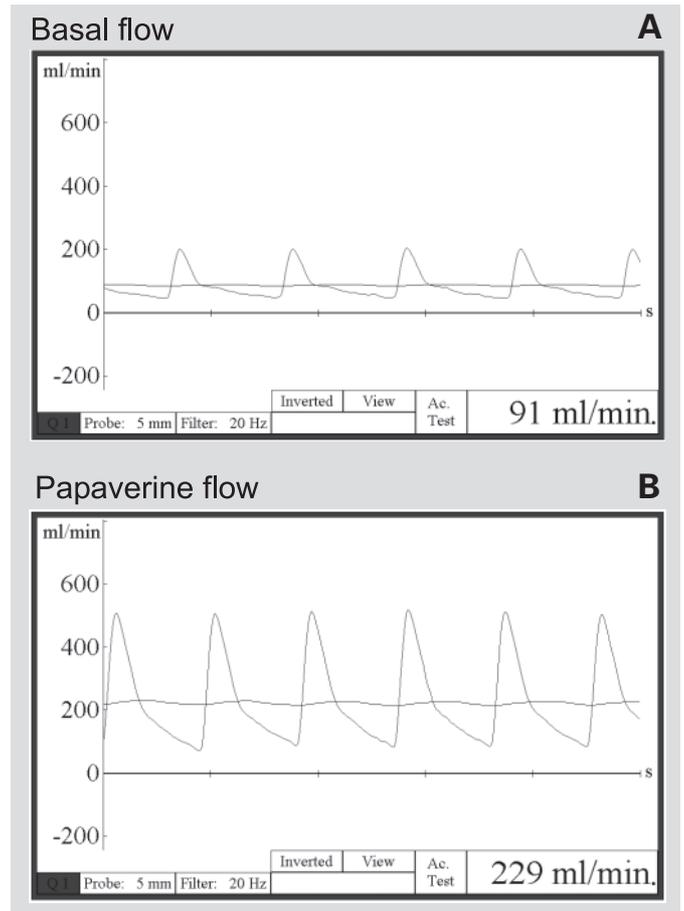


**Fig. 10.** Procedure for the detection of side branches during in situ bypass surgery. During the examination, the finger is compressing the artery. If net flow is detected (Pos. 2), the finger is downstream a side branch acting as arteriovenous fistula. This is subsequently ligated.

Obviously, blood flow values do not necessarily provide information about anatomical aberrations due to technical failure. Ideally, intraoperative arteriography or B-mode scanning is performed in order to supply the surgeon with an anatomical evaluation as well.

#### *Intra-Arterial Pressure Recording*

Arteriography alone may be inadequate in the prediction of the hemodynamic significance of lesions, particularly in the aortoiliac region, as reported by Sumner and Strandness [28]. The study noted that up to 30% of patients who had combined aortoiliac and femoropopliteal disease did not show hemodynamic improvement after a proximal reconstruction. This confirmed the inadequacy of arteriography alone in predicting the outcome of a proximal arterial reconstruction.



**Fig. 11.** Mean and pulsatile blood flow curves in a successful femoropopliteal bypass. **A** Basal flow. **B** Following intra-arterial injection of 40 mg papaverine.

Most investigators have generally agreed upon that any attempt to predict hemodynamic significance of a stenosis must be verified by direct intra-arterial pressure measurements [29]. Normally, the systolic pressure gradient between the aorta and the common femoral artery is below 10 mm Hg. In cases with unclear angiographic findings and a gradient below 10 mm Hg, intra-arterial papaverine is given, leading to a transient vasodilatation and increase in flow. Administration of papaverine is designed to mimic the flow increase seen with exercise. A systolic pressure gradient of more than 20 mm Hg is considered hemodynamically significant [30].

There is little doubt that combined intra-arterial pressure and arteriography is extremely useful for documenting the hemodynamic status of an arterial segment. It has been suggested that these measurements must be done when stenotic lesions are found and angioplasty is

being contemplated [29]. Furthermore, the pressure gradient should in addition be measured after completion of the dilatation to document the immediate result.

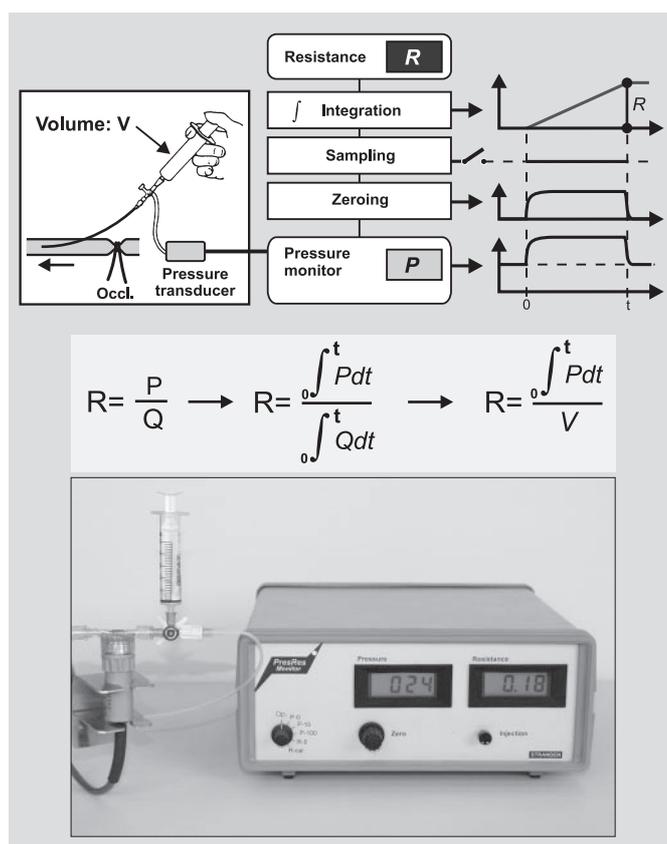
When reference pressure is obtained at the arm, great care should be taken when interpreting the data. Carter [31] and Carter and Tate [32] have shown that large systolic pressure wave amplification in the peripheral vessels may exist, especially in body cooling. Therefore, if brachial, or even worse, radial pressure is taken to represent central prestenotic pressure, an overestimation of that pressure may be significant. Hence the pressure gradient over the stenosis may be overestimated. This effect is reduced when the body is kept warm to induce peripheral vasodilatation.

When pressure is recorded during angiography, a 'pull-through' technique should be applied, whenever possible. The pressure transducer catheter line is introduced and moved proximal to the obstruction to measure the pre-obstruction pressure. Then the catheter is pulled through the obstruction to obtain the post-obstruction pressure. The difference between these pressures indicates the degree of obstruction.

#### Measurement of Vascular Outflow Resistance

Following femorodistal bypass surgery, both limb salvage rate and graft patency are dependent upon outflow resistance. Preoperative arteriographic visualization of these vessels does not inform us on whether their functional capacity is adequate to keep a distal bypass open. Studies have shown that intraoperative measurements of distal resistance are predictive in terms of outcome of femorodistal bypasses [33–35]. Some authors have defined a sharp cutoff point above which femorodistal grafts are prone to reocclude [34]. Attempts have been made to define a level of resistance above which primary amputation should be performed instead of arterial reconstruction [36], or levels where reopening of the graft can be neglected in case reocclusion occurs in the early postoperative phase. However, factors like the nature of the fluid infused during the measurement or the tone of the peripheral vascular tree, for example, may significantly influence the results [36, 37].

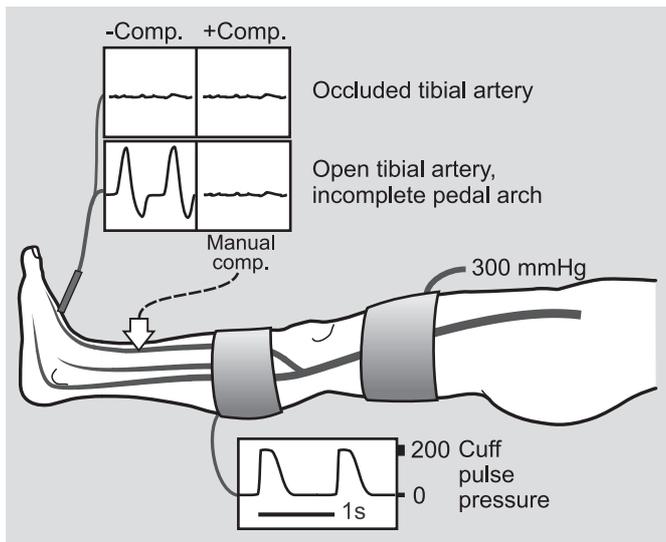
Outflow resistance (R) has been examined with different techniques, one of which is the basic *pressure (P)* divided by *flow (Q)* measurement, which represents the definition of vascular resistance:  $R = P/Q$ , expressed in peripheral resistance units ( $\text{mm Hg} \cdot \text{min} \cdot \text{ml}^{-1}$ ). This technique was simplified, as described by Ascer et al. [33]. A mathematical rewriting simplifies the equation to include the pressure integral when a fixed volume (V) is



**Fig. 12.** Outflow resistance measurement by the pressure integral technique. The mathematical basis is shown in the equation. Vascular resistance (R) is by definition the mean pressure (P) divided by mean blood flow (Q), for instance the pressure increase when a volume is injected by a syringe. This expression may be integrated as shown in the middle equation. However, as the time integral of flow is equal to the volume injected in that time period (V), the middle equation is reduced to constitute the pressure integral divided by a fixed volume. In the unit shown below ('PresRes Monitor', developed by the author) the vascular outflow resistance is obtained by pressure recording during infusion, subtracted of backpressure (zeroing), sampled during a fixed infusion time (infusion flow 100 ml/min) and automatically integrated during that time. When the proximal part of the artery is occluded ('Occl.'), the distal runoff resistance is recorded. When the occlusion is removed, some of the fluid may be squeezed also in proximal direction, and the total vascular resistance distal to a vascular 'end-to-side' graft is recorded.

injected into the distal artery (fig. 12). In this technique, flowmetry is not needed as a special pressure integration unit is used.

Noninvasive techniques examining outflow vessels preoperatively have also been presented. The most promising is the pulse-generated runoff, where a pressure cuff is placed around the calf or thigh, and Doppler examination of the calf (or foot) vessels is performed after rapid repeti-



**Fig. 13.** Schematic presentation of the noninvasive pulse-generated runoff technique for evaluating outflow vessels prior to surgery. A proximal occlusion cuff prevents interference from the proximal arterial pulse. A pressure cuff placed at calf level is intermittently inflated to produce artificial pressure/flow waves, which may be detected in the distal branches when the artery is open (lower left curve). The pedal arch is examined by manually compressing (+Comp.) the leg arteries as described by Scott et al. [40].

tive inflations of the cuff [38, 39]. The method has a semi-quantitative scoring system. With the technique, a standardized artificial arterial pressure wave is created in the calf and detected in the leg/foot arteries by Doppler ultrasound technique. The latest refinement of the procedure includes a pedal arch patency test (fig. 13). This test, applied to all three leg arteries, may predict the likelihood for a femorodistal bypass to remain open, and could be of help to select patients where a primary amputation should be considered [40].

## Anatomical Tests

### *Intraoperative Arteriography*

Intraoperative arteriography is regarded as the gold standard in the control of infrainguinal reconstruction, either by conventional arteriography or digital subtraction arteriography. It has been shown that technical defects may be observed in more than 20% of the cases [41]. Due to the direct visualization of the vessels, interpretation of the findings is generally easy. By detecting defects already during the operation, they can be corrected, thereby improving the patency rate. However,

there are a few disadvantages with the method. Usually only the graft and the distal anastomosis are visualized, which is unfortunate since technical defects may also be located at the proximal anastomosis. The arteriograms are normally taken in one plane only, and small defects may be hidden in the contrast material. The method includes injection of contrast material, which may lead to allergic reactions, and may be nephrotoxic.

### *Magnetic Resonance Angiography (MRA)*

For pre- and postoperative evaluations, MRA has become a noninvasive alternative to arteriography. This technique is advantageous as it does not require exposure to ionizing radiation and arterial catheterization. Different MRA techniques have been developed, with the latest, contrast-enhanced MRA being the most promising for the evaluation of the aortoiliac and infrainguinal vasculature. The tendency of overestimating the degree of stenosis may still be a problem, though.

### *Angioscopy*

Following the introduction of modern fiberoptic angioscopes, the graft and the anastomotic area can be visualized. Angioscopy may be used to detect residual valvular flaps, which may be retained following in situ vein bypass grafting (fig. 14). The method can also be used with endovascular closure of venous side branches, and in evaluating venous valves. In some areas, angioscopy is superior to arteriography. Neville et al. [42] found in an in vivo model that planar arteriography in one projection identified 60% of the arterial intimal flaps, while angioscopy and intravascular ultrasound visualized all of them. Comparing angioscopy with arteriography, the latter method is especially inferior in identifying thrombi; sometimes arteriography underestimates or misses mural or retained thrombi and debris [43].

A disadvantage of angioscopy is the lack of visualization of the outflow arteries including the pedal arch, whereas angiography visually includes these arteries. Furthermore, only anatomical details may be shown, and no impression of the peripheral vascular resistance is obtained. Sometimes clear visualization is difficult due to bleeding from side branches, and the method may be technically difficult to perform. The equipment is expensive, especially disposable angioscopes, and there may be difficulties with cleaning and sterilization.

### *Color-Flow Duplex (CD) Scanning*

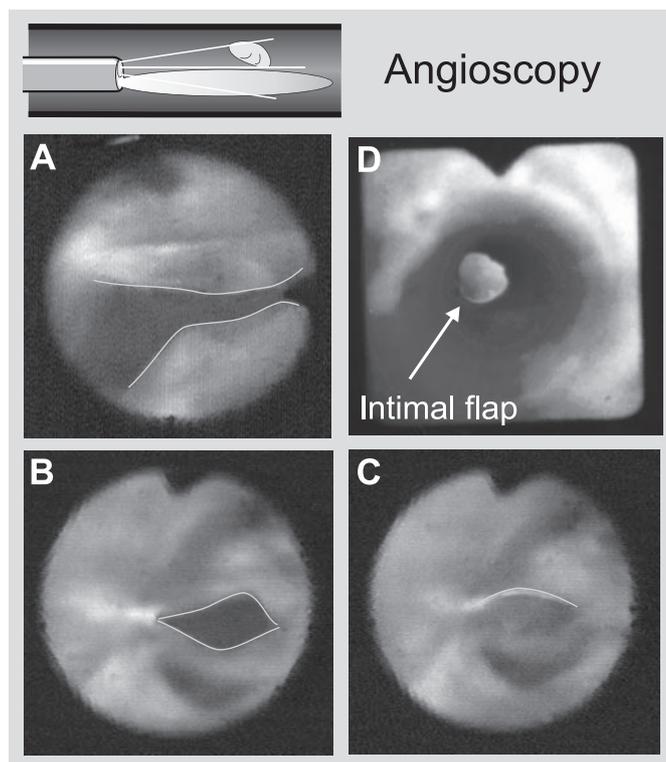
CD scanning is a direct noninvasive technique and includes three modalities: real-time *B-mode* (brightness

mode) image, *Doppler velocity spectrum* (usually PW) and *color-coded flow information* of blood velocity superimposed on the B-mode image. The technique provides both anatomical and physiological information, and has become the key instrument in vascular laboratories.

*Pre- and Postoperative Evaluation.* Color-coding has the advantage over plain duplex (B-mode + spectral Doppler) that it enables rapid identification of arteries, and specifically areas of stenoses and obstructions with flow disturbances. Atherosclerotic plaques are identified by the presence of calcification or increased echogenicity. Stenoses can be detected using different criteria, such as peak systolic velocity (PSV) within the narrowing, the ratio of PSV in the stenosis and PSV in a normal segment close to it (PSVR), end-diastolic velocity or, more subjectively, by noting the degree of spectral broadening. Several studies indicate a high degree of accuracy in the assessment of the aortoiliac and femoropopliteal arteries, but poorer accuracy in distal arteries [44]. Most centers use a PSV ratio  $>2.0$  as a criterion of stenosis to detect a stenosis with a diameter reduction  $>50\%$ . These studies have challenged arteriography as the golden standard in the evaluation of lower-extremity atherosclerosis, especially in the aortoiliac and femoropopliteal arteries. Consequently, an increasing number of hospitals now base the selection of therapy on CD investigations. For planning femorodistal reconstructions, most surgeons will find duplex scanning not optimal and regard arteriography a prerequisite.

Deep vessels, low blood velocities, calcified plaques with acoustic shadows and bowel gas frequently reduce the quality of the CD scanning, which is thus sometimes inconclusive. Software enhancement, for example power mode Doppler based on signal intensity rather than the Doppler shift, increases sensitivity, and may be used to improve locating vessels with low blood flow states or improve delineation of hypoechoic plaques. Furthermore, echo enhancers (ultrasound contrast), administered intravenously, may significantly improve the investigation in problem areas, e.g. in arteries of the pelvis, distal thigh (Hunter's canal) and calf.

Duplex scanning provides tissue images in multiple planes on the operating table, together with a velocity profile analysis for the evaluation of the hemodynamic results. Color codification of the blood velocity placed on the tissue image provides additional global, two-dimensional information on the flow patterns in the vessels, anastomoses and grafts [45]. Furthermore, anatomical defects, e.g. loose intimal flaps, strictures, intraluminal thrombi, loose debris, residual stenosis and misplaced



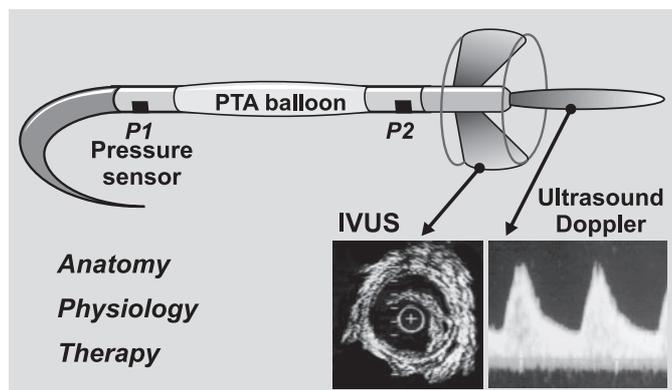
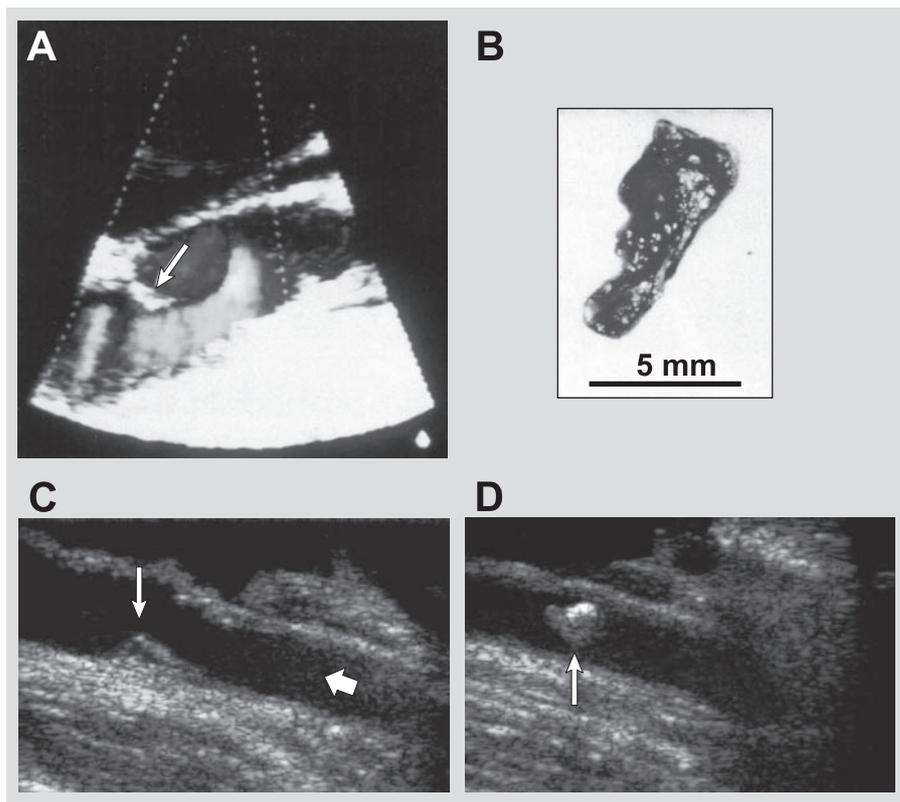
**Fig. 14.** Angioscopy images from venous (A–C) and arterial (D) investigations. **A** Incompetent venous valve with floppy cusps that do not approximate during Valsalva maneuver. The same valve after successful angioscopically guided external valvuloplasty during opening (**B**) and closing (**C**). The images are retained from a video sequence, hence the somewhat reduced image quality. For the benefit of clarity, the edges of the valve cusps are artificially outlined. **D** A residual intimal flap after arterial reconstruction.

sutures, may be identified [46] (fig. 15). Flow abnormalities found with Doppler spectral analysis or Doppler color-coding helps to identify significant lesions [47]. According to published reports on the intraoperative use of duplex ultrasound to assess reconstructive vascular surgery, defects are found in 20–30%. Only 5–10% of these are judged as significant and consequently corrected. The indications for vessel reentry are subjective and empirical. The decision to revise a reconstruction is principally based on the surgeon's judgment that the residual anatomical defect or hemodynamic disturbance may cause complications.

#### *Intravascular Ultrasound*

For endovascular procedures intravascular (intraluminal) ultrasound may be the method of choice, since it can visualize the composition of the lesion, and thereby help to select the proper treatment modality. Furthermore,

**Fig. 15.** An intimal defect (arrow) protruding into the lumen following carotid endarterectomy detected by color duplex scanning (A). At reintervention, an intimal flap of 2 × 7 mm was identified and removed (B). C B-mode image of a residual intimal flap detected following carotid endarterectomy (arrow). The wide arrow indicates the direction of blood flow. After revising the defect, a new examination identified a fresh thrombus (D, arrow) at the site where an arterial clamp was applied during the first revision. Consequently, a second reopening was necessary. The images are kindly provided by O.D. Saether (A, B) and T. Dahl (C, D), Trondheim University Hospital, Trondheim, Norway.



**Fig. 16.** Suggested single introduction, therapeutical IVUS unit with a PTA balloon that might be available in the future. This unit combines the description of intraluminal *anatomy* (IVUS), *physiology* [pressure gradient (P2-P1) and Doppler ultrasound] and *therapy* (PTA with/without stent) in one introduction procedure.

residual plaques or flaps can be detected, and a sufficient stent wall contact can be confirmed [48]. The technique will earn more attention when combined with *therapy constitution*. A single introduction, therapeutic IVUS unit with a PTA balloon, which could be reality in the future, is

hypothesized in figure 16. As microcatheter technology is being refined, one might expect that addition of dual pressure sensors and a Doppler ultrasound transducer at the tip should not raise the catheter cost significantly. With the pressure sensors, indicated as P1 and P2, the pressure gradient is measured continuously to evaluate the immediate effect of angioplasty. Before dilatation, the estimated degree of stenosis is however overestimated, as the catheter occupies a fraction of the cross-section area of the narrowed lumen. These measurement entities are already available on separate units: miniaturized pressure sensors are available in angiography guide wires (e.g. Cardiometrics WaveWire, Endosonics, San Diego, Calif., USA), as is Doppler ultrasound transducers (e.g. Cardiometrics Flo-Wire, Endosonics). Combined IVUS/PTA catheters are already available (e.g. MegaSonics F/X and OTW PTCA catheters, Endosonics).

#### *Three-Dimensional Mode*

Vascular disorders are essentially three-dimensional and should logically be visualized that way, especially in regard to vascular malformations and the quantification of atherosclerotic plaques [49, 50]. Current technology has been used for three-dimensional imaging of larger

organs like the heart and kidney, and volume estimation of abdominal organs [51, 52]. In particular, three-dimensional echocardiography has proved to render important additional clinical information [53–57].

### Three-/Four-Dimensional Representation

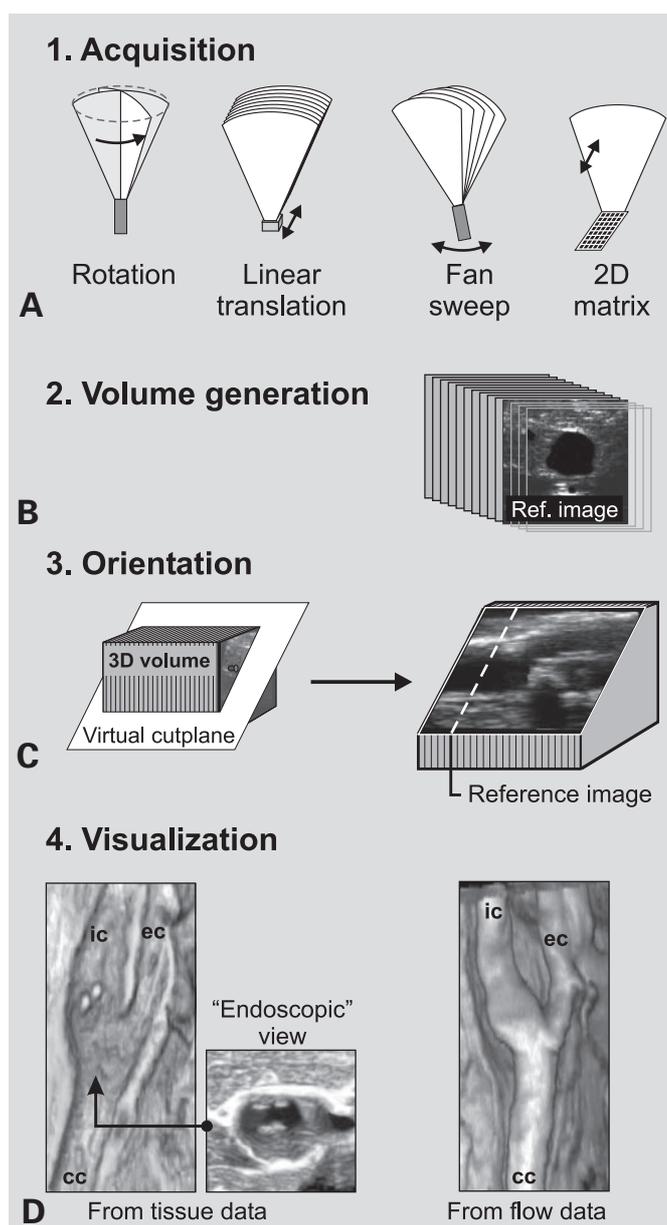
A two-dimensional ultrasound image is composed of X·Y picture elements, or *pixel*. This concept can be extended to the imaging of three-dimensional objects. The depth, or Z dimension, can be modeled by stacking a series of two-dimensional planes in depth. The two-dimensional picture elements are then interpolated to become cubes. These cubes, which form the three-dimensional image, are called volume elements or *voxels*. The volume representation is called a cuberille. When volumetric information is correlated with time to form the fourth dimension, every voxel element has variations over time. Representation in four dimensions is particularly used when time dynamics plays an essential role, e.g. displaying the heart or cardiovascular structures over a cardiac cycle.

At present, the process of making three-dimensional images based on ultrasonography is divided into four steps (fig. 17): data acquisition and filtering; volume generation; data processing, and visualization.

Several methods are available to acquire two-dimensional images to be transformed into three-dimensional data. The region of interest may be insonated by movement of the probe, by rotating, tilting or linear translation. When a position sensor system is applied, for instance based on magnetic tracking, the transducer may be moved about freely to generate the set of two-dimensional images that are transformed into the three-dimensional domain [58]. In the future, two-dimensional array transducers are probably employed to a larger degree. Very high processing speeds and capacities are however needed for real-time three-dimensional display. The acquisition may involve respiration and ECG triggering to reduce movement artifacts.

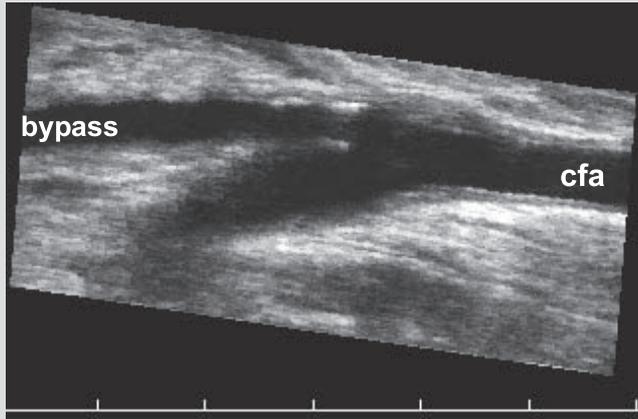
During three-dimensional data acquisition and image processing, the data undergo several transformation and filtering operations. The first post-processing step is to transform the two-dimensional video-grabbed data into a cubic representation of the image volume by interpolating neighboring pixels in the two-dimensional video-capture.

The next post-processing step, orientation, refers to the interactive process of reviewing the data when ‘viewing’ windows are opened for examining the volume. This viewing function is the essential tool for the positioning of viewing planes in relation to objects or structures that are of clinical interest.

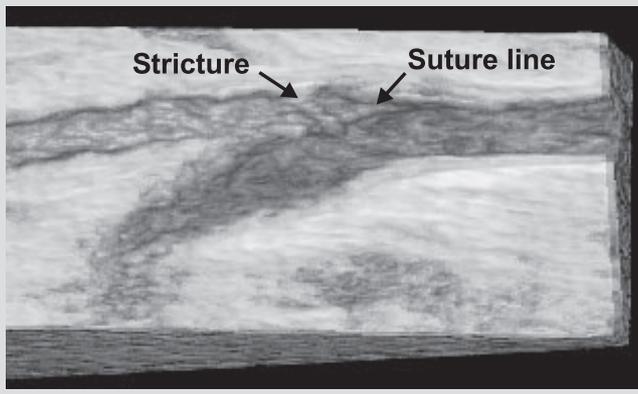


**Fig. 17.** Ultrasonic 3D acquisition (A) and post-processing (B–D). **A** A 3D image volume is created by translation of the ultrasound transducer, by rotation, linear translation or fan sweep. When a position sensor system is applied, the transducer may be moved about freely to generate a set of 2D images. In the future, 2D matrix transducers are used most probably. **B** Volume generation: a series of 2D images are placed one behind the other to form an image volume. A reference image is chosen within this series where important clinical information is displayed. **C, D** Visualization of the region of interest within the image volume is done with virtually cut planes (viewing planes) in any direction and location within the volume. These cut planes may be freely moved and rotated at any angle. The cut planes are slices in the image volume and visualized as images either based on tissue or flow data (D).

### 2D ultrasound data



### 3D ultrasound data



**Fig. 18.** Views of the proximal anastomosis of a patient with a reversed vein femoropopliteal bypass to the common femoral artery (cfa) applying the 3D gradient-shading technique. The 3D image is generated from the longitudinal reference image at the top and subsequently underlying parallel planes (in depth). The arrows point to the suture line of the anastomosis and what is supposed to be a wall stricture.

The final phase of post-processing is the computation of three-/four-dimensional images containing information on tissue surfaces, their orientation and distance from the viewing plane. Surface points that are more distant from the observer will become darker than points that are closer to the observer. Our studies on vascular three-dimensional processing [59] indicate that gradient shading is an applicable and robust technique. In gradient shading, the orientation of surface together with distance determines the gray value, providing images with visual appearance correlating well with normal photographic techniques. Figure 18 depicts the proximal anastomosis of a patient with a reversed vein femoropopliteal bypass applying the three-dimensional gradient-shading technique.

Color Doppler velocity or power energy mode data have been used to isolate vascular structures from gray scale tissue images, thereby obtaining clearer vessel anatomy [59–61]. The use of velocity or power mode data has a twofold advantage over gray-scale-generated images. Firstly, a major difficulty in volume data visualization is that the majority of volume elements do not contain clinically relevant data. Extraction of clinically useful information may therefore be difficult other than displaying a two-dimensional slice. Using flow information reduces this problem. Secondly, eliminating gray scale and instead using flow data for surface rendering also improves visualization of vascular structures (fig. 19). This is especially important when understanding complex spatial relationships.

Three-dimensional representation of the vasculature may become a valuable adjunct to conventional two-dimensional imaging. An advantage of the three-dimensional representation lies in the fact that orientations are preserved, enabling descriptive data, not definable by two-dimensional formats or a verbal report, to be conveyed rapidly and effectively to the clinician. This may enable less-experienced physicians to interpret complex images and to make diagnostic decisions that currently require specialists in sonography. The effectiveness of this transfer of information, however, depends on how realistic the three-dimensional images appear.

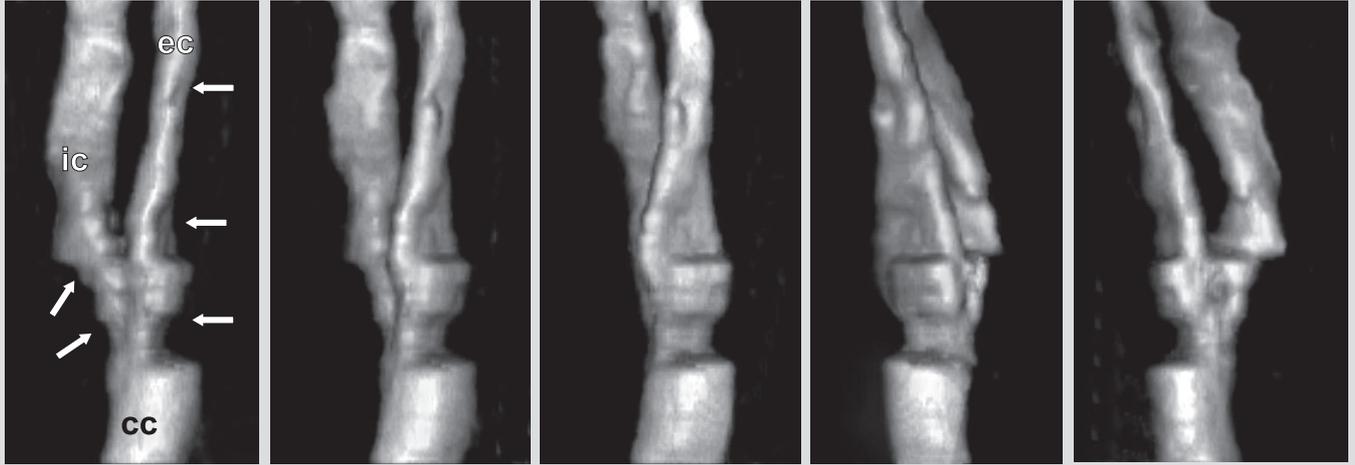
Mathematical modeling may be applied to the acquired dataset to calculate the volume of the structures [51]. The Vingmed EchoPac 3D ([www.geultrasound.com](http://www.geultrasound.com)) facilitates different acquisition modes and is computer platform independent. Interactive contouring does the volume modeling, and the volume calculations have proven very accurate. An example is shown in figure 20.

Our experience shows that three-/four-dimensional ultrasound is superior to two-dimensional imaging in communicating volumetric information, and the use of the ultrasound power energy mode may overcome problems related to poor B-mode visualization. Clinically, three-dimensional imaging may become important in relation to arterial reconstructions performed without angiography, e.g. carotid endarterectomies or even beating heart coronary surgery in the future.

#### *Cost-Effectiveness*

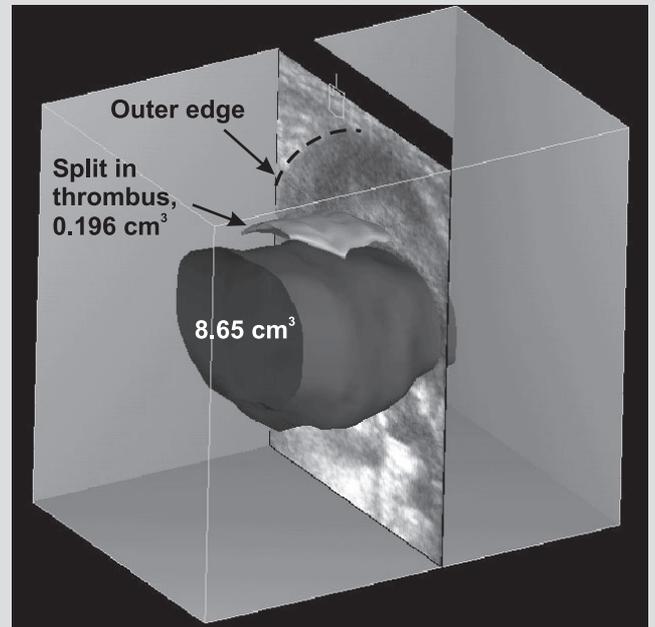
In the future, it is likely that health economy will be even more constrained than today. Cost-effectiveness of intraoperative methods will therefore have to be evaluated thoroughly before they are introduced into the clinical routine. The selection of methods will be based on the

### 3D ultrasound - rotation



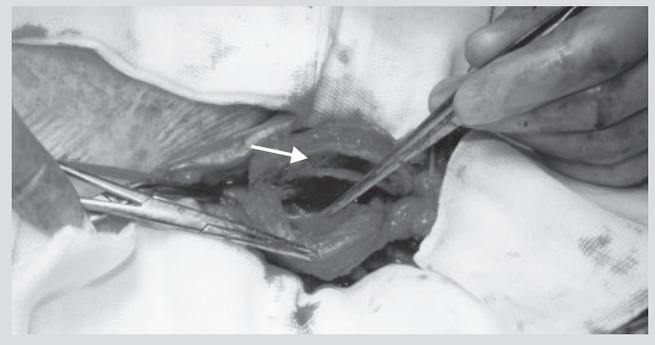
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### Femoropopliteal bypass aneurysm



**Fig. 19.** Sections from rotation of a 3D image of a carotid bifurcation. The images are based on power mode Doppler data. Tissue data are suppressed by reducing the tissue gain level. The arrows point to irregularities of the flow contour, coinciding with stenotic parts of the vessels. cc = Common carotid artery; ic = internal carotid artery; ec = external carotid artery.

**Fig. 20.** Calculation of the volume of a femoropopliteal bypass aneurysm using Vingmed Echopac 3D. The volume is the demarcation of the open flow section of the aneurysm. The volume outside this contour, underneath the vessel wall (interrupted line), is thrombotic material. The lighter section above the flow-generated volume is a split in the thrombus, verified at surgical removal of the aneurysm (arrow).



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number of reconstructions performed by the institution each year, and the economical situation of the hospital. Furthermore, the procedure must be easy to use, and should not prolong the operating time unduly.

The costs of the procedures must be weighed against the possibility of leaving a less satisfactory reconstruction. In the long run, this could lead to graft occlusion with amputation as a result in patients with critical limb ischemia [62, 63]. Initially the cost of amputation is similar to vascular reconstructions, but the patient with an amputation may be a burden for the society for many years. Consequently, efforts to improve the patency rate of reconstructions for critical limb ischemia can be regarded as good economy.

### Future

Angiography is one of the cornerstones in the intra-operative diagnostics in critical limb ischemia, and in many centers regarded as part of the minimum facilities in the management of these patients. Angioscopy still has to be evaluated as far as the efficiency of detecting significant failures are concerned. These methods will probably supplement each other in the near future.

Further refinements in three-dimensional imaging of the vessels are anticipated, especially in the ultrasound techniques. Therapeutic IVUS catheters with pressure transducers and Doppler ultrasound for PTA and stent insertion probably also emerge. Real-time three-dimensional ultrasound scanners require a fast computer kernel and improved transducer technology, probably a crystal

matrix design. With such technology, a three-dimensional representation of blood flow and flow vectors may enhance our perception of local flow conditions and morphological changes in diseased states. Along that chain of evolutions, we hope that color maps for energy dissipation are developed and implemented to directly locate and visualize the hemodynamic importance of arterial obstructions.

Resistance measurements will probably have a more important place in this area in the future. Patients who end up with an amputation following vascular surgery have often undergone several attempts of revascularization prior to amputation. Thus, a key problem is to decide when to do a primary amputation, or when to give the patient the chance of reconstruction, and when to refrain from a second attempt in case reocclusion occurs. Finally, resistance measurements can perhaps give us some idea when it will be feasible to perform a femorodistal bypass graft to the leg arteries. Reliable data concerning cutoff values are warranted.

Two basic functional tests are going to persist: flowmetry and intra-arterial pressure measurements. Of the flowmeters, transit time technology will be chosen because of its accuracy and ease of handling. When combined, these measurements may constitute outflow resistance.

### Acknowledgment

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