Review

Bio-acoustic signals from stenotic tube flow: state of the art and perspectives for future methodological development

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Abstract—*To study the degree of stenosis from the acoustic signal generated by the turbulent flow in a stenotic vessel, so-called phonoangiography was first suggested over 20 years ago. A reason for the limited use of the technique today may be that, in the early work, the theory of how to relate the spectrum of the acoustic signal to the degree of the stenosis was not clear. However, during the last decade, the theoretical basis for this and other biological tube flow applications has been clarified. Now there is also easy access to computers for frequency analysis. A further explanation for the limited diagnostic use of bio-acoustic techniques for tube flow is the strong competition from ultrasound Doppler techniques. In the future, however, applications may be expected in biological tube flow where the non-invasive, simple and inexpensive bio-acoustic techniques will have a definite role as a diagnostic method.*

Keywords—*Acoustic techniques, Blood flow, Obstruction, Phonoangiography, Turbulent flow, Urethra*

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1 Introduction

UNDER THE term bio-acoustic techniques we include different acoustic techniques used to study various biological phenomena. Among such techniques, the stethoscope used for auscultation is probably the oldest medical instrument. By replacing the sensing ear with a microphone, the phonocardiographic technique was established and has been used for decades. Recordings have also been used of acoustic signals generated from obstructed flow in vessels. Recently, a similar technique has been suggested to study urethral obstruction.

Acoustic methods to study biological fluid flow have the attractive feature of being non-invasive, and are usually simple and inexpensive. The phonocardiographic technique was used increasingly until the 1970s when it was, to a large extent, replaced by the ultrasound Doppler technique. However, acoustic and ultrasound techniques measure two completely different signals, as indicators of the biological flow; the sound and the flow velocity, respectively.

The aim of this work is to review the state of the art for acoustic signals as indicators of flow in biological tubes, excluding the well established phonocardiographic technique. Furthermore, from this literature survey we wish to assess the potential usefulness of future diagnostic techniques based on the measurements of these signals.

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2 Review of theory and methods

Fluid flow approaching an obstruction contracts in the entrance region (Fig. 1). The streamlines continue to converge beyond the entrance of the obstruction until they become parallel at a short distance from it (MERIT, 1967, BLEVINS, 1984). Here the jet flow reaches a minimal area at the vena contracta. At the vena contracta, the pressure reaches a local minimum. As the flow downstream of the vena contracta diverges, the local velocity decreases and pressure recovery occurs.

The flow regime is judged from the Reynolds number

$$Re = \rho du/\eta \qquad (1)$$

where ρ is the density of the fluid, d is the diameter (or more generally the characteristic length), u is the average velocity and η is the viscosity.

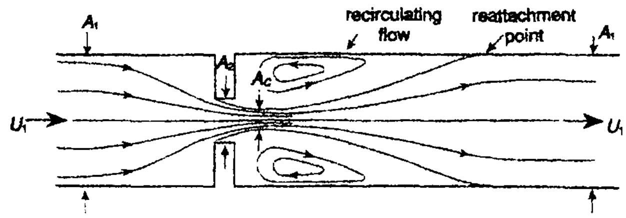

Fig. 1 *Streamlines of flow through and distal to stenosis; A_1 = area proximal and distal to obstruction; A_2 = area at obstruction; A_c = area of vena contracta; streamlines leave the wall proximal to the orifice and reattach at a certain distance distally*

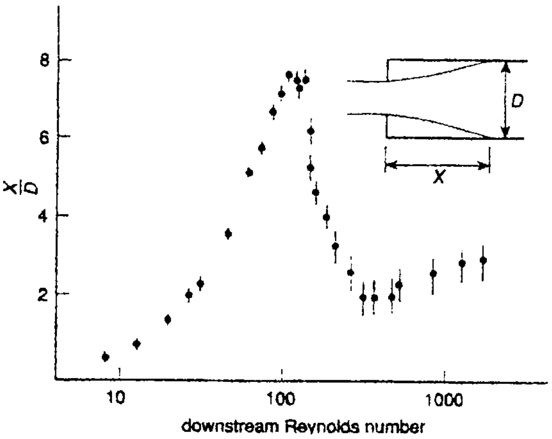

Fig. 2 *Location of reattachment point normalised to downstream diameter D shown as a function of the downstream Reynolds number; measurements were made for a 85% stenosis; D = 24·8 mm (drawn after BACK and ROSCHKE, 1972)*

For an upstream Reynolds number lower than 1, the fluid follows the tube boundaries even at the outlet of the obstruction and the pressure gradient is low (MASSEY, 1979). However, when the Reynolds number increases, up to about ten separations occur. This separation produces a jet where the shear layer between the jet and the surrounding fluid becomes unstable. With increasing flow, vortices are formed which are shed downstream. Downstream from the orifice, the jet diverges and, at a certain site, it makes contact with the tube wall. This point is called the reattachment point.

At relatively low Reynolds numbers (10–100), the point of reattachment moves downstream as the laminar shear layers grow (Fig. 2) (BACK and ROSCHKE, 1972; TOBIN and CHANG, 1976). For higher Reynolds numbers, the reattachment point moves backwards towards the stenosis. This latter phenomenon is associated with instability in the shear layer. For Reynolds numbers greater than about 325, the shear layer becomes highly disturbed. The reattachment point in this case is close to the orifice and it is only gradually displaced downstream of the hole as the flow rate increases.

The turbulent flow consists of velocity fluctuations that are superimposed on the main velocity. Each change in velocity causes changes in force, and hence pressure variations. The root mean square (RMS) value of the fluctuations gives a quantitative measure of the turbulence intensity (MASSEY, 1979), which generally increases as the Reynolds number increases (KIRKEEIDE *et al.*, 1977). The pressure variations affect the tube wall and cause it to vibrate in the acoustic frequency range.

Burns performed an analysis of murmurs in the cardiovascular system and explained them by the vortex streets known to be generated in a free field distal to an obstruction (BURNS, 1959). The frequency of the vortices depends on the geometry of the obstruction. Burns suggested that the frequency of the vortices could be determined from the Strouhal number.

The observation that obstructions in the carotid vessels may cause cerebral damage was the incentive to study the sound generated in stenotic vessels. Kartchner and McRae measured sound generated by flow by placing a microphone over the cervical carotid artery (KARTCHNER and MCRAE, 1969; 1973). The recorded signals were evaluated qualitatively from the tracing on an oscilloscope, and the technique was found to be an accurate diagnostic technique in extra cranial carotid occlusive disease.

Lees and Dewey were the first to present the principles of quantitative phonoangiography (LEES and DEWEY, 1970). Using a piezoelectric displacement transducer placed on the skin, they measured the acoustic signal generated by turbulent flow in larger stenotic arteries. They found that the frequency range of interest of phonoangiographic signals is 50–1000 Hz. Spectra recorded in carotid stenosis showed a peak and a roll off in the intensity above a break frequency (DUNCAN *et al.*, 1975). The obtained frequency spectrum of the measured signal was similar to that produced by turbulent flow in pipes.

Other investigators (DUNCAN *et al.*, 1975; KISTLER *et al.*, 1978; MILLER *et al.*, 1980a,b; KISTLER *et al.*, 1981; LEES, 1984) have studied the relationship between pressure fluctuations recorded on the skin wall and the size of the stenosis. The degree of stenosis in the vessel, expressed as the diameter at the occlusion, was calculated from the quotient between the peak systolic flow velocity in the unoccluded part of the vessel and the break frequency of the spectrum. Kistler *et al.* suggested the following relation with a Strouhal number (KISTLER *et al.*, 1978):

$$S_1 = \frac{f_b d}{u_u} \qquad (2)$$

where f_b is the break frequency, d is the diameter of the stenosis and u_u is the linear flow velocity in the unobstructed part of the tube.

An uncertainty in obtaining the diameter of the obstruction from the Strouhal number S_1 was the estimation of the flow velocity in the tube. An empirical relation was, however, determined between this velocity and the intensity of the acoustic signal (DUNCAN *et al.*, 1975).

Fredberg developed a theoretical model for how the pressure fluctuations due to turbulence in the vessel can be observed as sound on the skin surface (FREDBERG, 1974). With this model, he could show that the spectrum of the pressure variations at the wall is related to the spectrum of the pressure signal on the skin via a filtering factor approximately proportional to $1/f^2$, where f is the frequency of the pressure signal. The explanation for this filtering effect was not the absorption or damping of the signal but the way stochastic signals are added at the surface.

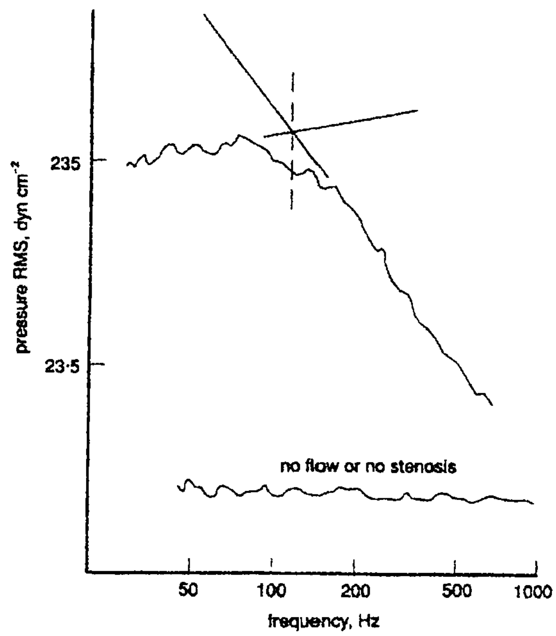

Fig. 3 *Spectrum of wall pressure recorded in experiment modelling stenosis in tube; position of the pressure measurement at maximal RMS value; spectrum is compared with the spectrum at no stenosis; break or corner frequency is identified (drawn after TOBIN and CHANG, 1976)*

An advantage of the phonoangiographic technique is that it is non-invasive, relatively inexpensive and easy to perform, involves no risk to the patient, and the investigation can be performed with minimal human and machine power. The original analysis of the problem by Lees and Dewey involved both the frequency and intensity of the recorded signal for the determination of the obstruction (LEES and DEWEY, 1970). The involvement of the intensity in the calculations complicates the estimation of the vessel width, because the acoustic signal is damped, to a varying degree, when propagated from the vessel through the tissue to the transducer head. Furthermore, with phonoangiography the identification of a 'break frequency' in the frequency spectrum may present difficulties. Manual identification of this frequency may have poor reproducibility, and accurate computer algorithms for identification are not readily available.

Flow velocity downstream of a stenosis was measured by Kim and Corcoran by the use of a hot-film anemometer (KIM and CORCORAN, 1974). Studies were performed for a circular orifice in a rigid-wall tube. They obtained smooth frequency spectra without frequency peaks which have been predicted by other authors (BURNS, 1959). They therefore suggested that earlier relations developed for free jets were not valid for their confined jet. The series of vortices generated distal to the orifice, which if maintained should correspond to a peak frequency in the velocity spectrum, are therefore presumably becoming more randomised due to reattachment to the wall and mixing with the surrounding fluid, for example.

Kim and Corcoran also studied sound measured from a contact microphone on the surface of latex tubes in which plexiglass obstructions were placed. Obtained spectra were dependent on the geometry of the stenosis and the properties of the elastic tube material. The spectra of the velocity and sound signal showed different characteristics.

Tobin and Chang performed model experiment wall pressure measurements (TOBIN and CHANG, 1976). The frequency spectrum of the wall pressure was studied for different degrees of obstruction and at different locations from the stenosis. In the obtained frequency spectra, break or corner frequencies f_b were identified (Fig. 3). For a specific geometry, they defined a Strouhal number

$$S_2 = \frac{f_b d}{u_j} \tag{3}$$

where d is the diameter of stenotic orifice and u_j is the flow velocity of the jet. Note that in this Strouhal number the flow velocity in the obstruction is used instead of the flow velocity in the unobstructed part of the vessel, as in the previous expression (S_1 in eqn. 2). A model experiment verified the Strouhal number relationship of eqn. 3. Tobin and Chang therefore suggested this number as a tool to predict the severity of stenosis in humans.

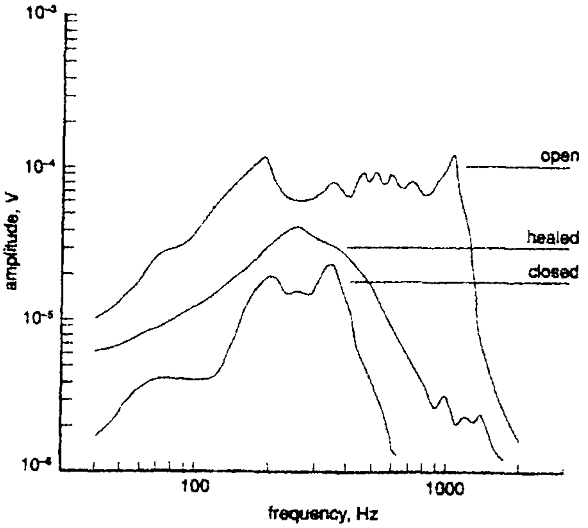

Fig. 5 *Frequency spectra of bruits recorded from exposed carotid artery in dog (open), from skin surface after wound closure (closed) and 12 weeks later (healed); frequency peak in upper part of spectrum disappears when tissue surrounds vessel; frequency peak with characteristic roll off is obtained when wound is healed (drawn after MILLER et al., 1980b)*

Miller *et al.* studied the Strouhal number relationship in experiments on dogs. (MILLER *et al.*, 1980a,b). They found that the relationship between the break frequency, flow velocity and residual vessel lumen diameter was valid over a wide range of values (Fig. 4). Their investigations also showed that the acoustic signal transmitted to the skin surface was strongly dependent on the tethering of the vessel (Fig. 5). With a free dissected vessel, resonance peaks were obtained from oscillations in the vessel wall. However, with the vessel in its normal place, the resonances are damped out by the surrounding tissue and a smooth spectrum is obtained. The contribution from the resonances are superimposed on the smooth spectrum. The spectrum associated with the resonances in the vessel wall is called the resonance spectrum. The smooth spectrum is assumed to be caused by the flow turbulence and is therefore called the turbulence spectrum.

Clark studied models of aortic stenosis and measured velocity and pressure spectra for different orifice geometries (CLARK, 1976; 1977). He found that the normalised power spectra of the velocity depended on the downstream Reynolds number, but only to some extent on nozzle size and geometry. The normalised spectrum of pressure, on the other hand, did not depend on the downstream Reynolds number, but on the orifice geometry.

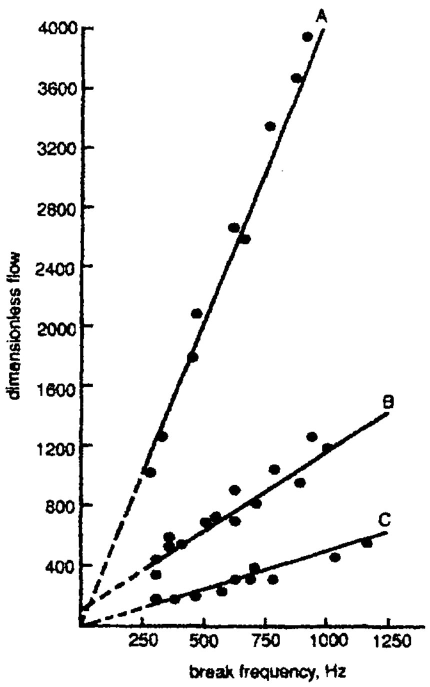

Fig. 4 *Relationship between break frequency and relative volume flow measured in 11 dogs; A, B and C = relations for increasing severity of stenosis (drawn after MILLER et al., 1980a)*

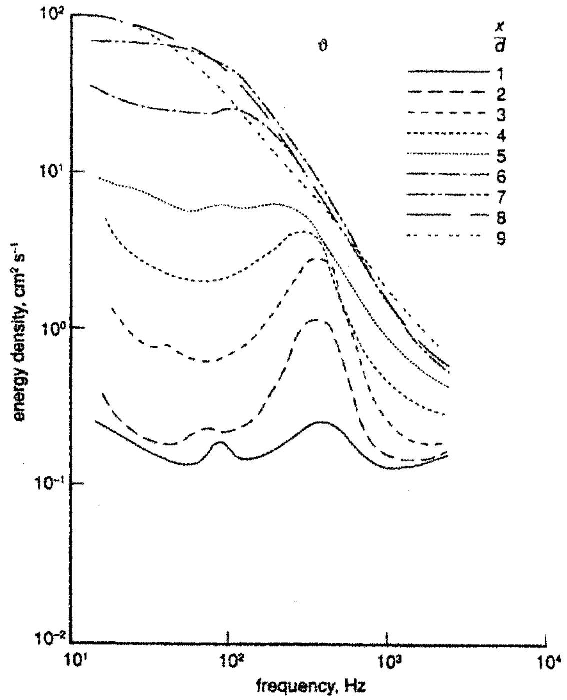

Fig. 6 *Frequency spectrum of axial component of flow velocity measured in tube at different distances from nozzle (x) and expressed as number of nozzle diameters (d); velocities are measured along centre axis; close to nozzle frequency, peaks can be observed corresponding to well organised vortex shedding; further away from nozzle smooth spectra are obtained (drawn after LU et al., 1983)*

Studying turbulent pulsatile flow in an orifice jet, Cox *et al.* found velocity signal spectra that depended on the orifice size (COX *et al.*, 1979). In model experiments, Fredberg found that the pressure fluctuations from a nozzle jet were related to the kinetic energy of the jet for Reynolds numbers greater than 2000, but for lower Reynolds numbers the amplitude of the pressure fluctuations depended on both the velocity of the jet and the viscosity of the fluid (FREDBERG, 1977).

In an animal experiment, Lu *et al.* measured pressure fluctuations distal to a surgically created stenosis (LU *et al.*, 1980). The frequency spectrum of the flow velocity signal showed a power slope of $-5/3$ and then reached a break frequency, after which the slope changed to $-10/3$.

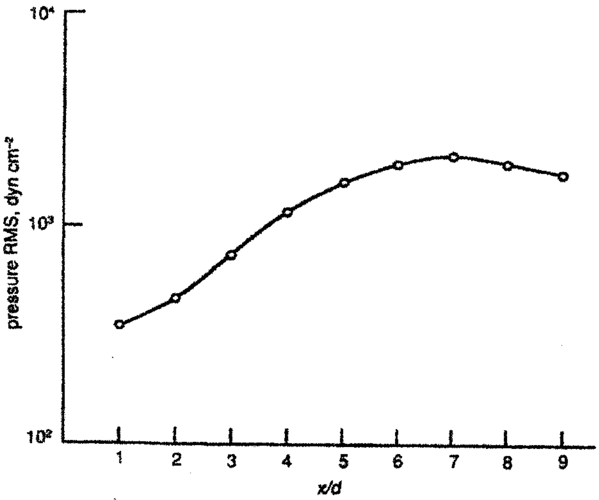

Fig. 7 *RMS value of pressure fluctuations at wall of tube shown as function of distance from orifice, normalised as distance x over orifice diameter d (drawn after LU et al., 1983)*

In a plexiglass model experiment, LU *et al.* measured two components' flow velocity using laser Doppler anemometry and pressure using a piezoelectric device distal to the stenosis. The characteristic frequencies of the two velocity components in the mixing layer were generally different. Near the nozzle, the velocity spectra showed narrow peaks, which were expected to be related to the organised periodicity of the vortex shedding in this area (Fig. 6). These peaks disappeared, however, as the distance downstream from the nozzle was increased. The energy of the periodic velocity fluctuations close to the orifice was much smaller than that of components further downstream.

The RMS of the pressure fluctuations recorded at the wall is shown in Fig. 7 as a function of the distance from the nozzle. The maximal intensity is obtained at a distance of seven nozzle diameters distal to the hole.

The peak frequency of the pressure spectrum (Fig. 8) is different to that of the velocity components, and this difference increases the closer to the hole the measurements are performed. The dependence of the wall pressure spectra on the distance from the hole is also shown. However, this dependence vanishes for a distance greater than about seven orifice diameters. The spectrum increased proportionally with f for frequencies up to a break frequency and as $f^{-1.6}$ for larger frequencies (LU *et al.*, 1983).

Jones and Fronek pointed out the difficulties in objectively determining the break frequency of a spectrum (JONES and FRONEK, 1987) and suggested that this should be determined from the second moment of the spectrum as

$$f_b = \left(\frac{\int_0^{f^2} f^2 P(f) df}{\int_0^{f^2} P(f) df} \right)^{1/2} \tag{4}$$

Furthermore, they investigated what type of Strouhal number relationship should be used for determination of the stenotic diameter. In a model experiment, they measured flow velocity with a hot film probe. They found the Strouhal number S_2 (eqn. 3) correlated better with their data than the S_1 number (eqn. 2). From their data, an improved relation was suggested:

$$S_2 = Re^{0.72}(d/D)^{0.26} \tag{5}$$

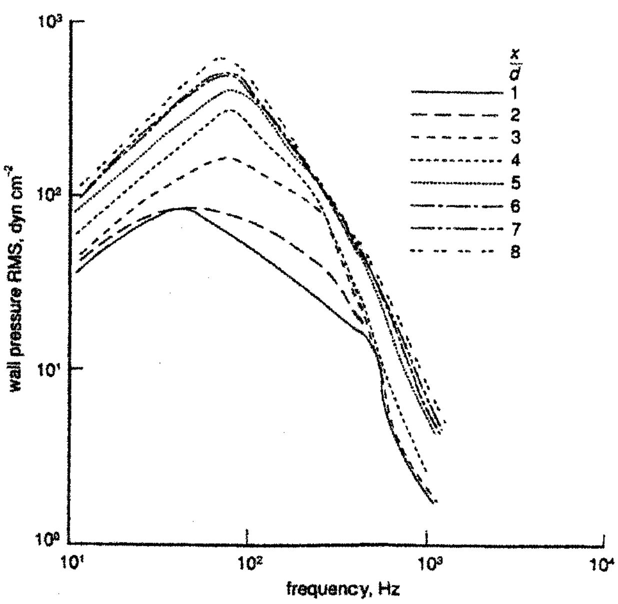

Fig. 8 *Frequency spectrum of pressure at tube wall shown for different normalised distances (x/d) from orifice (drawn after LU et al., 1983)*

where D is the diameter of the unobstructed part of the tube. It should be mentioned that this empirical relation is based on intraluminal velocity measurements. From the work by Lu *et al.* (LU *et al.*, 1983), the break frequency estimated from pressure measurements on the tube wall or skin surface is not normally the same as that of the velocity spectra. The Strouhal relation derived empirically from pressure data is therefore expected to be different.

Abdullah and Hwang have studied, both theoretically and in model experiments, the relationship between the velocity fluctuations of the flow and pressure at the wall (ABDULLAH and HWANG, 1988). As the shear noise of the flow downstream to the obstruction is known to be the major contributor to the pressure fluctuations at the wall near the reattachment point, the frequency distribution of these fields was studied in various parts of the flow field. The normalised power spectral density of wall pressure at the reattachment point for different Reynolds numbers and from different studies showed a consistent relationship with a break frequency at a certain Strouhal number (Fig. 9). Correlation studies between the flow velocity at different locations in the flow field and the wall pressure suggested that vortices shed from the jet area with a certain frequency are the major cause of the pressure fluctuations at the wall reattachment point. The spectrum of the pressure fluctuations here is related to the jet Strouhal number S_2 (eqn. 3) and does therefore reflect conditions at the obstruction. Further downstream the initial condition at the jet is lost and flow parameters are determined only by local parameters of the flow.

Wang *et al.* developed a model for sound generated from stenosis in coronary arteries (WANG *et al.*, 1990). The coronary artery was represented by an analogue electrical network. The obtained transfer function showed frequency characteristics with two resonance peaks. The height of the frequency peak was found to contain information on the degree of stenosis, and the location of the peak was altered as the degree of obstruction was changed in the model. The simulated transfer function was similar to the frequency spectrum of the acoustic signal recorded in a patient with coronary artery stenosis (Fig. 10). The resonance peak of the highest frequency was present when first recorded but disappeared after angioplasty.

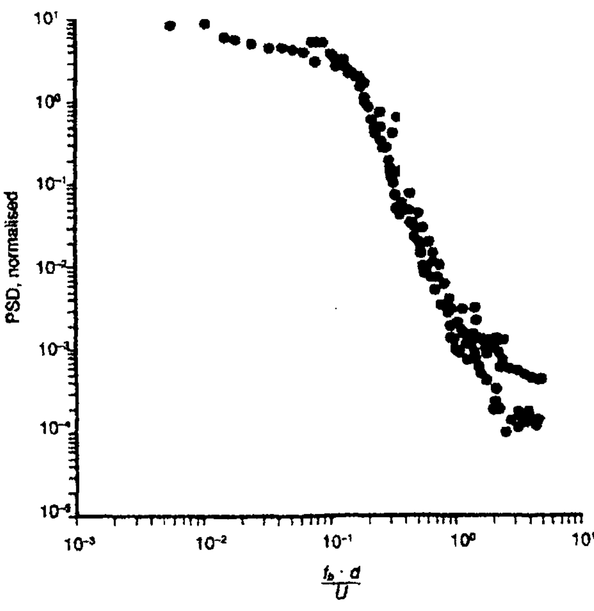

Fig. 9 *Normalised powers spectral density as function of Strouhal number (S_2, eqn. 3); plot is made for different Reynolds numbers and from different studies (drawn after ABDULLAH and HWANG, 1988)*

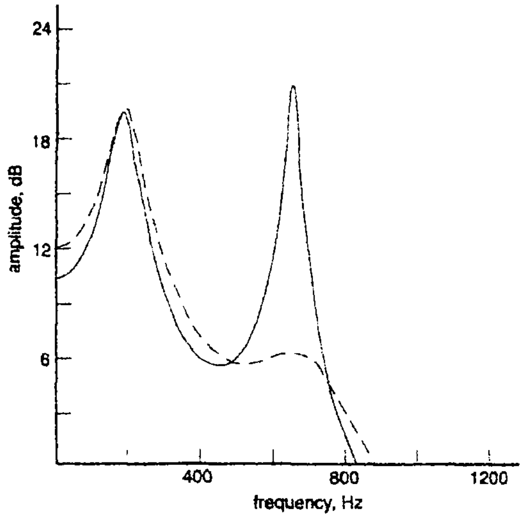

Fig. 10 *Frequency spectrum of acoustic signals from coronary artery shown in patient before angioplasty (solid line) and after (dashed line); higher frequency resonance peak disappears after removing stenosis (drawn after WANG et al., 1990)*

We have previously suggested a method similar to the phonoangiographic technique to assess the degree of obstruction of urethral urinary flow in urethral obstruction (TERIÖ, 1989; 1991). The method was studied in a model experiment, where the urethra was represented by silicone tubes with internal diameters of 3·4–6·4 mm placed in a water-filled test chamber. Slit-type obstructions with smooth edges were obtained by compressing the tube. The acoustic signal was obtained from a microphone placed close to the obstruction in the chamber. An example of the recorded signal spectra from this study is shown in Fig. 11, in which the increasing higher frequency content for obstructed flow can be seen.

From the recorded spectra, the mean power frequency was calculated. Fig. 12 shows a unique relationship between the quotient of flow over the width of the obstruction and the mean power frequency. This indicates therefore that, if the volume flow is known, the degree of obstruction can be estimated. Examples of recorded spectra from patients are shown in Fig. 13.

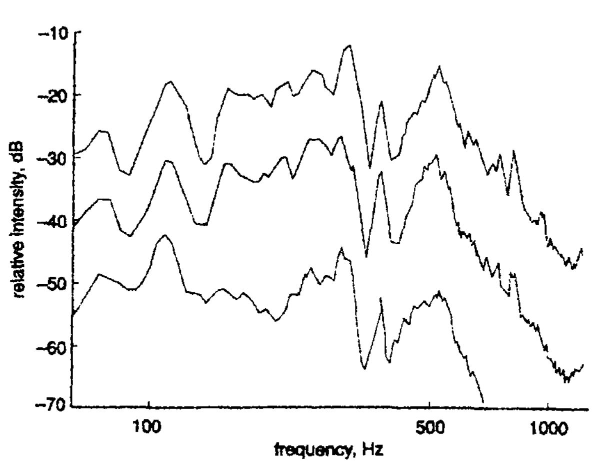

Fig. 11 *Frequency spectra recorded 50 mm distal to obstruction in urethral model consisting of elastic tube with unobstructed diameter of 3·4 mm; curve 1 = no obstruction; curve 2 = obstruction width 1·6 mm; curve 3 = obstruction width 1·0 mm (reproduced with permission from TERIÖ, 1991)*

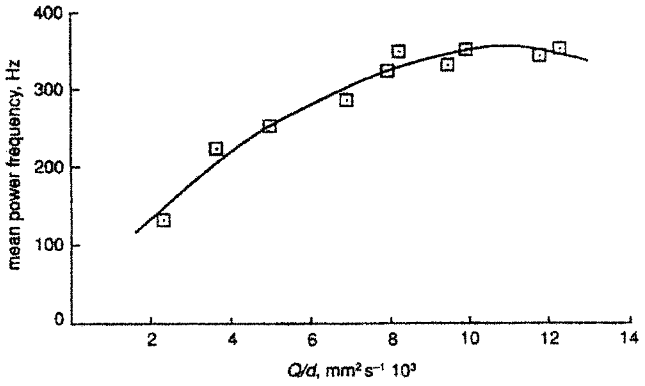

Fig. 12. *Mean power frequency as function of relationship between volume flow Q and diameter d recorded 50 mm distal to obstruction in urethral model (reproduced with permission from* TERIÖ, *1991)*

3 Discussion

The use of bio-acoustic techniques for studying the degree of stenosis in biological tubes goes back for more than 20 years. During this time, the technique has not succeeded in establishing itself as a clinically accepted method for quantifying stenosis. The reasons for this could be the limitations of the technique and the availability of other powerful investigation techniques, which have definite advantages over the bio-acoustic methods.

Angiography remains an established technique for the investigation of the presence of stenosis in blood vessels. An advantage of this technique is that information about the location of the stenosis is obtained simultaneously with the image of its size. The information from angiography about the stenosis remains, however, qualitative or semi-quantitative because the cross-sectional area estimation involves gross errors from projection errors and poor image resolution. Furthermore, an angiographic investigation involves expensive equipment and specialised staff, and it is an invasive investigation not without risk for the patient.

The non-invasive alternative to the bio-acoustic technique is the use of Doppler ultrasound. This method is by far the most common technique when investigating vessel stenosis and is the method clinically used nowadays instead of phonoangiography. The ultrasound Doppler technique can provide both quantitative and qualitative information on the flow. Duplex ultrasound instruments have the advantage of producing both tissue and flow 2-D imaging.

Recently, flow studies with magnetic resonance imaging (MRI) have become possible with commercial equipment which enables the investigation of biological tube flow. However, as this is a very complicated and expensive technique, it is not really appropriate to compare this technique with the bio-acoustic techniques. The indications for the MRI technique are for specialist investigations, whereas the bioacoustic technique can be performed even at the level of primary healthcare.

One reason for the poor acceptance of the described bio-acoustic technique may be that it is difficult to determine the relationship between the flow in the stenosis and the pressure measured at the skin wall. The theoretical background is quite complicated. In the early work in this field, the observed pressure parameters were mainly empirically related to the size of the stenosis. Over the years, different theories have been presented for the pressure generation and different Strouhal numbers have been suggested.

From this review, it can be concluded that the theoretical background has been strengthened over the years and that currently a solid theoretical base exists which relates the degree of stenosis to the pressure fluctuations at the wall. The shear flow between the jet and the surrounding fluid at the stenosis generates vortices that are shed downstream. These vortices cause pressure fluctuations at the vessel wall. The pressure at the wall reaches its maximum at the reattachment point of the jet, generally located about seven orifice diameters downstream from the hole. The pressure here also has the strongest correlation to the eddy flow at the stenosis. The measurements of the bio-acoustic signal should therefore be taken as close to this location as possible. From recorded pressure frequency spectra, stenosis characteristics can be calculated from a Strouhal number S_2 (eqn. 3). It is also possible that this relationship can be made more sophisticated.

An early obstacle for the wider use of the phonoangiographic technique was the need for frequency analysis and other signal-processing tasks. At the time, this meant that expensive instrumentation had to be acquired; nowadays this type of analysis can be performed on an inexpensive computer.

It can be difficult to identify precisely the break frequency of the spectrum used in the Strouhal number calculation or for the calculation of other flow-related parameters. One method of identifying the low- and high-frequency displacement of the spectrum is to calculate the momentum of the spectrum (JONES and FRONEK, 1987; TERIÖ. 1989; 1991), which is easily performed in conjunction with Fourier anlaysis on a computer. A definite advantage with the latter method is that all the information in the spectrum is used in the calculation, instead of just the part involving the break point. This makes it a more robust measure.

Returning to the usefulness of the bio-acoustic technique, comparison should be made with the single beam ultrasound Doppler technique. Principally, the bio-acoustic technique has the advantage of being passive, in the sense that energy does not have to be conveyed from the transducer to the measurement object. This implies that, because the bio-acoustic technique does not include the transmission process, we do not have to worry about disturbances when transmitting the signal from the transducer to the measurement object or about the need for scatterers that ensure the reflection.

The ultrasound beam has to be aligned with the tube flow, which is sometimes not possible due to obstacles; whereas the bio-acoustic signal is irradiated in a broad angle. The ultrasound and the acoustic sound are also attenuated quite differently. Bone and air are definite obstacles for the

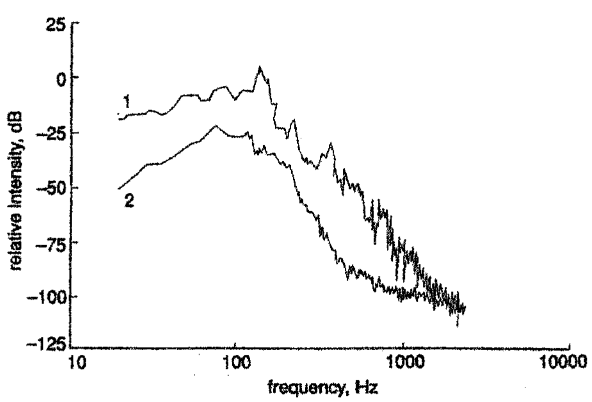

Fig. 13. *Frequency spectra of acoustic signal recorded from urethra in normal man (10 mm dorsally from the scrotum); curve 1 = induced obstruction; curve 2 = spectrum from unobstructed urinary flow (reproduced with permission from* TERIÖ, *1991)*

ultrasound, whereas these can be circumvented by the acoustic signal. On the other hand, the bio-acoustic signal is, however, disturbed by surrounding noise.

The bio-acoustic technique is basically a very simple and inexpensive technique. Compared with the ultrasound technique, acoustic signals are less damped and do not require scatters in the fluid. However, in the investigation of obstructions in easily accessible blood vessels, the ultrasound technique has the advantage of measuring the flow velocity directly. However, at locations where the propagation of the ultrasound is obstructed by bones, for example, or where the backscattered signal is reduced due to the absence of scatterers, as in urine, the bio-acoustic technique should have a role. Furthermore, the strengthened theoretical description of the signal generation presented in recent years has provided a basis for a sound application of the method. The problem of practical measurements in humans is that of disturbances from background noise. New methodological developments may solve this problem.

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