Continuous recording of pulmonary artery pressure

Sir,

The statement by Ikram and his coworkers in their paper on continuous pulmonary artery pressure recording (1984; 51: 421-6) that: "The total (measurement) system has a frequency response linear to 8 Hz allowing accurate pressure recording over the full range of heart rates" should not pass unchallenged.

Ikram suggests that, although the catheter-transducer pressure measurement system is underdamped with its resonant peak at 16 Hz, these characteristics are compensated by the limited frequency response of the Medilog tape recorder and playback system. The problems of low frequency catheter resonance have been discussed in detail by McDonald¹ and by Gabe.² Gabe pointed out that if the catheter-transducer damping factor is around 0.2, which my measurements on a portable transducer perfusion unit and No 5 Goodale Lubin catheter confirm, the highest frequency component which can be measured without distortion of the signal is about 20% that of the resonant frequency; in this case up to 4 Hz. Ikram presents a Millar catheter record of pulmonary artery pressure, a simultaneous record obtained by the transducer catheter system, which shows resonance, and a "clean" version of the record after replay through the recording system (Figs. 3 and 4). Although the waveforms are said to match each other, the pulse

pressure is about 15% greater in the underdamped system compared with the Millar catheter, and after replay the pulse pressure is apparently attenuated by nearly 50%. Furthermore, the gross distortion of the replayed "clean" waveform in the region of the dichrotic notch leaves one with little confidence that the systolic and diastolic pressures will be recorded accurately.

Ikram shows the frequency response of the entire recording system (Fig. 2 in his paper). Using the measured frequency response of the Medilog recorder and replay unit (AM4 recorder amplifier and PB2 replay unit with PM3 amplifiers) and by treating the catheter and transducer as a classical second order system with its variables determined by a "pop" test,¹ the frequency response of the entire system may be approximated and is shown in Fig. 1. The agreement with Ikram's results is reasonable, although the -2 dB (21%) attenuation at 2.5 Hz shown in his paper cannot be reproduced precisely. When this curve is plotted as transmission rather than gain in decibels it is clear that the response of the system is far from flat within the physiologically significant range of frequencies.

This passband characteristic arises from three effects. The signal is attenuated by a simple presettable filter in the Medilog replay system, and attenuation will occur throughout the physiologically significant range of frequencies. This attenuation is, however, offset with increasing frequency because the effect of the resonant peak of the catheter becomes more significant. Also, the demodulator in the replay unit uses a very sharp filter with a corner frequency set at 8 Hz. Although this filter has a nominally flat passband, there is a slight increase in transmission at around 8 Hz (this overshoot in transmission is adjustable using a preset control in the replay unit) so further offsetting the attenuation of the simple presettable filter.

Such a system does not damp catheter resonance correctly. It relies on the slightly non-ideal characteristics of the replay unit demodulating filter which varies between individual units. It also relies on precise catheter resonance characteristics. McDonald¹ and Gabe² emphasise that the frequency response of a catheter transducer system near its resonance depends on the exact resonant frequency and the damping. The frequency response of the catheter, and hence the response shown in Fig. 1, may be dramatically modified by the formation of small bubbles within the catheter, which lower the resonant frequency and

Fig. 1 Plot of gain in decibels (unbroken line), transmission (○) and phase (▲) as a function of log₁₀ (frequency) for the transducer-catheter system and the Medilog recorder. These curves were obtained from the measured frequency response of the recorder-playback system (filter 4, gain 5) and the calculated characteristics of the catheter-transducer using a "pop" test.

Fig. 2 Mean power spectral density of replayed signals from the Medilog recorder after recording a 16 Hz tone (unbroken line) and 60 Hz (symbols). The replayed signal (filter 0, gain 5) was digitised at 12 kHz with the 3 dB point of the antialiasing filter set at 6 kHz. The signal was recorded in blocks of 256 points, tapered with a half cosine bell, and the power spectrum obtained using the FFT algorithm. The curves are the means of 43 records.

increase the damping, and also by the formation of very small clots at its tip, which increase the damping.

The shape of a recorded signal depends not only on the relative amplitude of the various Fourier components but also on their respective phases. An ideal system, such as an optimally damped resonant system, has a phase response which increases linearly with frequency and corresponds to delay in the signal without modification of its shape. The calculated phase response of the catheter, recorder, and replay system is shown in Fig. 1. It is decidedly non-linear in the passband, resulting in severe signal distortion.

The expected error in pulse pressure using this system can be estimated by convolving a known simulated waveform (derived from Millar catheter data) and the measurement system impulse response. This calculation shows that there can be up to $\pm 15\%$ error in measured pulse pressure which will depend on the heart rate, replay unit filter setting, and the shape of

the measured waveform. In practice, the overall behaviour of the recording system is likely to be worse than is suggested in Fig. 1 because of intrinsic recorder noise. Sayers *et al* have shown that systemic arterial pressure waveforms, measured with a portable transducer perfusion unit and a Medilog recorder, are degraded so that they cannot be distinguished from noise above 6 Hz.³

Finally, Ikram's frequency response plot shows two resonant peaks, one at 16 Hz and the other at about 60 Hz, which is attributed to the NIH catheter. The 60 Hz peak is very heavily attenuated by the applied filter and the resonant mode at 16 Hz to approximately -90 dB. The amplitude of this peak would therefore be expected to be in the region of 5000. There is, however, no possibility of this peak being a genuine part of the measured signal because the Medilog recorder uses pulse width modulation to record a DC biased signal and has a carrier frequency of 40 Hz.⁴ A 16 Hz sine wave was recorded on a Medilog recorder and the smoothed power spectrum of the replayed signal estimated to see if the peak at 60 Hz represents the 16 Hz peak reflected in the first modulation sideband of the signal. In addition, a 60 Hz sine wave was recorded and the smoothed power spectrum of the replayed signal was obtained. The results of these experiments are shown in Fig. 2. The 16 Hz tone is replayed faithfully, but there is no evidence of a 60 Hz component representing a modulation sideband. The 60 Hz tone is undetectable on replay.

The data presented by Ikram and his coworkers, and that obtained by myself, suggest that the combination of a portable transducer perfusion unit, a No 5 Goodale Lubin catheter, and a Medilog recorder in its DC mode has a frequency response which is inadequate for precise dynamic measurements of pulmonary artery pressure but is adequate for recording slowly moving fluctuations in mean pressure.

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References

- 1 McDonald DA. The design of manometers. In: *Blood flow in arteries*. 2nd ed. London: Edward Arnold, 1974: 174-208.
- 2 Gabe IT. Pressure measurement in experimental physiology. In: Bergel DH, ed. *Cardiovascular fluid dynamics*, vol 1. London and New York: Academic Press, 1972: 11-50.
- 3 Sayers BMCA, Ellis NE, Green H. Minimum and maximum requirements for physiological measurement; intra

arterial blood pressure. EEC contract report 297-76-12 ECI. Section 5. 1978.

4 Medilog MR-10 recorder service manual. Oxford: Oxford Medical Systems.

This letter was shown to the authors, who reply as follows:

Sir,

We would like to respond to Dr Saumarez's criticism of our technique for recording ambulatory pulmonary arterial pressure in man.

Our first comment is that he has assumed that Fig. 3 and Fig. 4 (p 424) come from the same patient. This is not the case and nowhere is this stated or implied in our paper. The purpose of Fig. 3 was to depict a comparison of waveforms from a high fidelity transducer tipped Millar catheter and a simultaneously recorded pressure using the Akers strain gauge with the fluid filled catheters used in our ambulatory system. Figure 4 is an example of the waveform produced by the Medilog playback unit, but they are from different patients. Hence his conclusions on waveform distortion based on that assumption are probably incorrect.

His comments on the frequency response of the total system are based on actual measurements of only some of the components, while the response of others has been calculated from published specifications. We, on the other hand, have tested the response of the entire system with a swept frequency generator and

Fig. 1 Phase response in relation to frequency response.

Fig. 2 Pulmonary arterial pressure (mm Hg) (a) at rest and (b) at submaximal exercise.

pressure chamber. This has produced the results depicted in our Fig. 2 (p 423).

His comments on the probable effects of micro-bubbles and clots on resonance and frequency response would apply to any catheter manometer system. In practice we have not found this to be a problem once a suitable catheter, heparin dose, and pump speed were perfected.

As regards the phase linearity, Saumarez's Fig. 2 plots phase response against the logarithm of the frequency. If phase is replotted against a linear frequency scale then it is evident that phase response is linear to above 7 Hz (Fig. 1).

Milnor states that 95% of the energy in the pulmonary arterial waveform lies within the first six harmonics.¹ This would suggest that our system is adequate to record the waveform at rest and moderate exercise. Figure 2 shows the waveform obtained in a patient at rest and during submaximal levels of exercise. This clearly shows systolic and diastolic peaks as well as the dicrotic notch and "a" wave at a heart rate of 120 beats/min. Despite theoretical problems, the system in practice does better than "follow slow oscillations around the mean" as Saumarez says. His reference to Sayers *et al's* comment² that systemic arterial pressure is unrecordable over 6 Hz using the Medilog system is inappropriate in the context of the pulmonary artery. In this vessel the mean pressure is low and the pulse pressure is a greater proportion of the mean pressure—that is the harmonics have a higher power content.

For these reasons we believe that our method is capable of recording systolic and diastolic pressures with reasonable accuracy but, as is readily acknowledged in our paper, it is incapable of high fidelity waveform reproduction.

The ultimate reason for developing the technique is to provide hitherto unobtainable physiological data on

the pulmonary circulation. Our studies on patients suggest that this is possible despite the obvious compromises inherent in a fluid filled low fidelity system. But until the inevitable development of better methods, it remains a practical method currently available for this purpose.

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References

- 1 Milnor WR. Pulmonary haemodynamics. In: Bergel DH, ed. *Cardiovascular fluid dynamics*, vol. 2. London and New York: Academic Press, 1972:299-340.
- 2 Sayers BMcA, Ellis NE, Green H. Minimum and maximum requirements for physiological measurement; intra arterial blood pressure. EEC contract report 297-76-12 ECI. Section 5. 1978.

This letter was shown to Dr Saumarez, who replies as follows:

Sir,

Ikram *et al* (p 583) have replotted the phase response in Fig. 1 of my first letter (p 581), and the replotted data purport to show that the phase response of their measurement system is linear over its operating range. They have not plotted the data correctly and the conclusion they have drawn is false. Figure 1 in my letter is a Bode plot. The characteristic of an ideal system when its phase is plotted in this way is that there is a constant phase increment for a constant frequency increment rather than that the plot is a straight line.

Admittedly, this is hard to see when the plot is reduced in scale. I have replotted the phase shift of the system as difference from the ideal linear phase change over the range 1-8 Hz (Figure), and it is quite clear that the phase response is non-linear. There is approximately 25° phase lead at 5 Hz. This has a profound effect on signal distortion. At a heart rate of 100 beats/minute, the second and third harmonics of the pressure waveform will contribute about 30-40% of their maximum amplitudes to the measured instantaneous pressure signal at a time when they have no contribution to the undistorted waveform. It is this effect, as well as amplitude distortion, which renders the catheter/transducer/Medilog system an unsatisfactory instrument for measuring any dynamic feature of the pulmonary artery pressure waveform.

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Figure Difference between the phase response of the catheter/transducer/Medilog system and the ideal linear phase response over the range 1-8 Hz plotted as a function of frequency.