Automating the Determination of Wave Speed Using the PU-Loop Method

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Abstract

The PU-loop (pressure-velocity loop) is a method for determining wave speed and relies on the linear relationship between the pressure and velocity in the absence of reflected waves. This linearity of the PU-loop during early systole, which is directly related to wave speed, has always been established by eye. This paper presents a new technique that establishes this linearity and thus determining wave speed online.

Pressure and flow were measured in the ascending aorta of 11 anesthetised dogs. The slope of the PU-loop, indicating wave speed was determined by eye and by using the new technique. The difference between the slopes of the two methods is in the order of 3%.

The new technique is convenient and allows for the online assessment of wave speed, which could be used as a bedside tool for the assessment of arterial compliance.

1. Introduction

Wave speed measurements can generally be grouped under two main categories; spatial and local wave speed. Spatial wave speed in the arterial system has been predominantly obtained using the foot-to-foot method which involves measuring either pressure or flow waveforms at two sites that are at a known distance apart, L. Dividing L by the temporal time that took the wave to run from one site to the other, Δt, gives wave speed. This method has been used in the clinical setting to determine arterial compliance [1] and the results confirms non-uniform wave speed along the aorta; a significant increase is observed distally [2]. Although this method is being used in the setting of patients care, the non-invasive results are somewhat controversial and subjective because of the difficulty in determining the foot of the wave and because of the assumption made about the distance between the two measurement sites due to the curvature of the arteries [3].

Local wave speed refers to the determination of wave speed at the measurement site. Westerhof *et al.* used the characteristic impedance to give an estimate of local wave speed [4]. Khir et al. have introduced the PU-loop method for determining local wave speed [5]. The method is based on the water hammer equation which describes that if waves at the measurement site are running in one direction, the relationship between pressure and velocity is linear. During early systole it is most probable that the only waves in the ascending aorta are those running forward. Therefore, the slope of the initial linear part of the PU-loop is directly related to wave speed, and has been conventionally determined by eye, i.e. a straight line was drawn over the initial linear part of the PU-loop, which was determined by eye.

In this work we present a new technique to examine the automatic construction of the slope of the initial linear part of the PU-loop.

2. Methods

Experiments were performed in 11 mongrel dogs (average weight 22 ± 3 kg, 7 males), anaesthetised with sodium pentobarbital, 30 mg/kg-body weight intravenously. A maintenance dose of 75 mg/hr was given intravenously for the duration of the experiment. The dog was endotracheally intubated and mechanically ventilated using a constant-volume ventilator (Model 607, Harvard Apparatus Company, Millis, Mass., USA). After a median sternotomy, an ultrasonic flow probe (Transonic Systems Inc., Ithaca, NY, USA) was mounted around the ascending aorta approximately 1 cm distal to the aortic valve. ECG leads were connected to both forelegs and the left back leg. A high-fidelity pressure catheter (Millar Instruments Inc., Houston, Texas, USA) was used to measure the pressure in the aortic root as near as possible to the site of the flow probe without creating interference (a few millimetres away from the flow probe, proximal to the aortic valve). The catheter was advanced from either the right or the left brachial artery.

Snares were placed at 4 different sites during the preparation of each dog: the upper descending thoracic aorta at the level of the aortic valve (thoracic); the lower thoracic aorta at the level of the diaphragm (diaphragm); the abdominal aorta between the renal arteries (abdominal) and the left iliac artery, 2 cm downstream from the aorta-iliac bifurcation (iliac). For each occlusion, data were collected for 30 seconds before the occlusion (control) and during the occlusion; 3 minutes after the snare was applied [6]. At each site, total occlusion was confirmed by observing no flow distal to the occlusion site, and another Millar high-fidelity pressure catheter was advanced from either the right iliac artery to measure pressure at the occlusion site. A time interval of 10-15 minutes was allowed between occlusions for returning to control conditions. In order to eliminate time effects, the sequence of occlusions was varied from dog to dog using a 4 by 4 Latin-square.

The circumference of the *post-mortem* ascending aorta was measured to convert the measured flow rate into velocity. We note that the circumference of the ascending aorta of each dog was measured at zero transmural pressure and hence the calculated diameter may be less than the actual diameter *in vivo*. However, in order to compensate for that difference, we did not take the wall thickness into account and assumed that the measured external radius is the correct value to be used in calculating the velocities.

The pressure catheters were calibrated prior to each experiment against a mercury manometer. Because of the possible time lag attributable to the filter in the ultrasonic flow meter, the foot of the pressure and velocity waveforms were aligned at the onset of ejection, and the lag was accounted for prior to carrying out the analysis. All *in vivo* data were recorded at a sampling rate of 200 Hz, stored digitally and were analysed using Matlab software (The MathWorks Inc., Natick, Mass, USA).

3. Analysis

The PU-loop method for determining wave speed is based on the water hammer equation, which can be written for the forward and backward waves as

$$\mathbf{dP_{\pm} = \pm \rho c dU_{\pm}} \qquad (1)$$

where dP and dU are the pressure and velocity differences over one sampling interval, ρ is blood density (1040 kg/m3) and c is wave speed. Equation (1) describes the relationship between the pressure and velocity which is linear if the waves are running in one direction. During early systole, probably the only waves in the ascending aorta are those running in the forward direction. Therefore, plotting the measured pressure against the measured velocity over the cycle we obtain a PU-loop, whose slope during the very early part of systole equals ρc as shown in Figure 1. On arrival of the reflected waves, the linear relationship between pressure and velocity will no longer hold and there will be a deflection point, after which the loop becomes non-linear. The algorithm described below is incorporated into a computer program that is written in MatLab (MathWorks Inc, Natick, Mass, USA) to automate the determination of the linear part in the PU-loop.

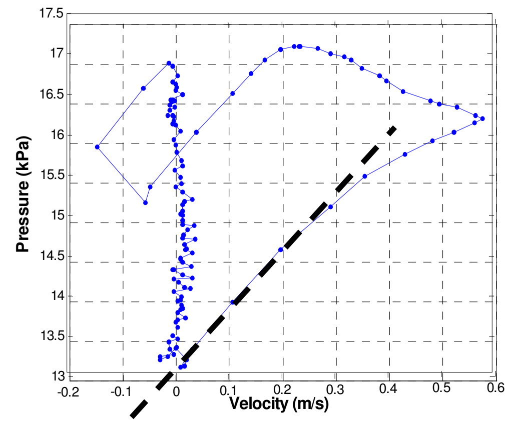


Figure 1. The PU-loop measured in the ascending aorta of dog at control conditions. The initial linear part of the loop, indicating wave speed of 6.6 m/s, is determined by eye and shown in the dashed line.

For a given PU-loop the program searches for the beginning of the linear part by calculating the average of *n* slopes following the slope being analysed, and the relative difference between this average and the current slope, is found using

$$\Delta S(i) = \frac{S(i)}{\left(\frac{1}{n}\sum_{i+1}^{i+n} S(i)\right)} - 1 \qquad (2)$$

where *S(i)* is the slope of the change in pressure (dP) and velocity (dU) over one sampling interval.

$$S(i) = \frac{P_{i+1} - P_i}{U_{i+1} - U_i} \qquad (3)$$

The relative difference, ΔS(i), is then compared with a tolerance level, τ, which is determined by trial-and-error and found to be of value of 0.35. This algorithm is a point-to-point technique, i.e. the program analyses one interval at a time, and keeps on processing the following intervals until it has found the beginning of the linear part, point K on Figure 2. This is established when

$$\Delta S(i) \le \tau \qquad (4)$$

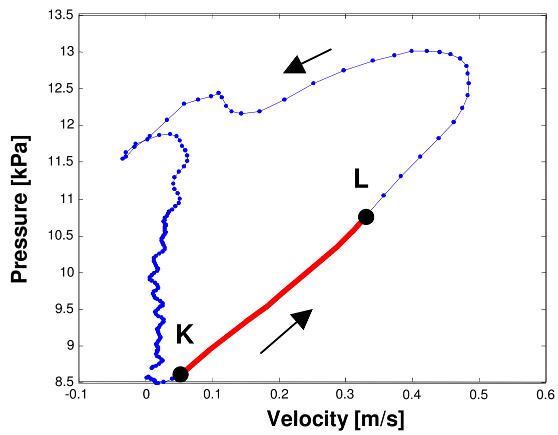
To find the end of the initial linear part of the PU-loop, the program calculates the relative difference between the current slope and the average of all the previous slopes starting from the beginning of the linear part, point K.

$$\Delta S(i) = \frac{S(i)}{\left(\frac{1}{i-K}\sum_{i=K}^{i-1} S(i)\right)} - 1 \qquad (5)$$

The program continues to compare $\Delta S(i)$ of Equation 5 to the tolerance level, τ , and the end of the linear part, point L on Figure 2, is found when Equation 4 is true.

4. Results

Figure 2a and 2b show the linear part of the PU-loop as determined by the new algorithm at control conditions and during abdominal occlusion.



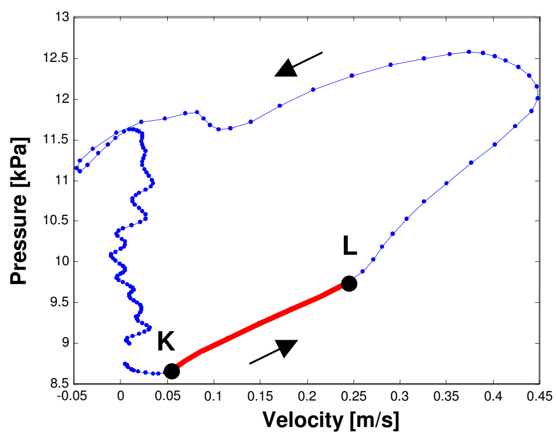


Figure 2. The PU-loop measured in the ascending aorta of dog at control conditions (a), and during abdominal occlusion (b). The initial linear part of the loop is determined using the new algorithm and shown in bold. The arrows indicate the direction of the loop. Sampling points (K) and (L) are the beginning and end of the initial linear part respectively. Although the end of the initial linear part of the PU-loop during control is less obvious than that during occlusion, the new technique accurately detects the length of the linear part in both cases.

5. Discussion and conclusions

5.1. The algorithm

The tolerance level τ, as discussed in the analysis, determines the accuracy of the linear part and represents the minimum allowed relative difference between slopes. We initially tested the algorithm and found by try and error that τ= 0.35 is the value that gave the best results. Then we tested the algorithm on a different set of data that was collected at 500 Hz and found the same value of τ has also given the best results. We therefore believe that the tolerance level is sample rate independent.

To establish linearity we decided that a minimum number of samples (n=4) should comply with the tolerance level as explained in Equation 4. Also, to make sure the program does not detect a linear part within the noise level; it calculates the length of the linear part and compares this length with the minimum number of samples. If the length of the linear part is below this minimum, the program assumes the found linear part is within the noise level and continues searching for the real linear part. Therefore, the program was written to detect the sample rate and adapt this minimum automatically. For example, our data was collected at 200 Hz, and the minimum length of the linear part of the PU-loop was 0.02 seconds.

As explained in the analysis, for detecting the beginning of the linear part, the program calculates the average of n slopes ahead of the slope being tested and determines the relative difference between this average and the current slope. Similar to the way we determined the minimum length of the linear portion of the loop, we found that when the slope being tested was compared with the average of n=4 slopes, yielded the best results in determining the beginning of the linear portion.

The tolerance level is a constant and sample rate independent. However the minimum number of samples (n) for locating the beginning and end of the linear part is sample rate dependant, can be determined by the program, and points (K) & (L) shown in Figure 2 can be determined accordingly. This procedure is a point-to-point technique, and the algorithm can be used to analyse immediately after it has collected the appropriate required number of samples, n. Wave speed can then be determined online.

The augmentation index (AIx) is used extensively in clinical studies to determine the magnitude of the reflected waves [7]. The accuracy of the technique relies on the ability to detect an inflection point on the pressure waveform upon the arrival of the reflected waves, using some derivative of the waveform. In our experiments a change in the pressure waveform was not always obvious. A comparison between the results of the AIx using the inflection point and using point L on the PU-loop has not been reported, and it is a question for a separate study.

5.2. Wave speed

Wave speed is an important property of the arterial segment, and it has been associated with pathological cardiac events [8]. It has also been suggested as a surrogate marker for aortic stiffness, which has been thought of as an independent predictor for cardiovascular mortality in hypertensive patients [9]. Thus, automating the determination of wave speed is of clinical relevance and the results of this paper are encouraging towards achieving this aim.

The PU-loop method is easy to use and requires the simultaneous measurements of pressure and flow velocity at the same site. However, because of the difficulty in obtaining accurate measurements of pressure non-invasively, other researchers have used a wall tracking system which allows for substituting pressure with vessel diameter [10], and used a similar online technique for determining wave speed based on wave intensity analysis. Harada *et al*. suggested that towards end of systole there maybe another period where waves in the ascending aorta are also unidirectional, similar to the period at early systole. We anticipate the relationship between pressure and velocity at that time to also be linear, and if desired, the computer program implementing the new algorithm can be adapted to detect other linear parts in the *PU*-loop.

The wave speed determined by using the PU-loop is traditionally made by establishing the slope of the initial part of the loop by eye. The new algorithm allows for automating this process and the average difference between the results of wave speed determined by eye and that determined automatically of is shown in Table 1.

	Wave speed [m/s]		
Dog no.	New algorithm	By eye	Difference [%]
1	4.67	4.60	1.6
2	8.70	8.18	6.4
3	4.25	4.22	0.8
4	4.83	4.82	0.1
5	4.38	4.23	3.4
6	9.65	9.20	4.9
7	8.03	7.79	3.0
8	5.52	5.42	2.0
9	5.87	5.91	-0.7
10	5.60	5.53	1.2
12	7.01	6.85	2.2
Average	**6.23**	**6.07**	**2.6**

Table 1: A comparison between wave speed determined by eye, and that determined using the algorithm described in this paper. Wave speed value of each dog, for each method is the average of all wave speeds measured at all of the interventions. The global average of the percentage difference between the two techniques is 2.6%.

We conclude that the PU-loop is an easy to use and mathematically sound method for determining wave speed. The new algorithm utilising the PU-loop is independent of sampling rate and the results compare well with those measured conventionally by eye. The new algorithm allows for the dynamic determination of wave speed online, which may be used as a bedside technique for the assessment of arterial compliance.

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