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AN EFFECTIVE ALGORITHM FOR MEASURING DIASTOLIC ARTERY DIAMETERS

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The assessment of the distensibility or compliance of an artery by means of ultrasound involves the determination of the instaneous change in diameter as well as the initial diamter. The change in diameter as function of time (distension waveform) can be assessed accurately using sample volumes tracking the positions of the vessel wall-lumen boundaries. However, manual positioning of these sample volumes poses specific problems related to the pulsatile behavior of the lumen diameter. Intermediate storage of rf data over a few seconds will eliminate these problems but this will limit recording time and will increase hardware complexity. A method is described to detect the wall-lumen interface synchronously with ECG allowing for automatic positioning of the sample volumes. The user interference is restricted to a rough identification of the lumen position using the envelope of the received signal. Starting from within the lumen the sample volumes are forced outward until the local envelope exceeds a threshold dynamically adjusting to the peak value of the nearby vessel wall signals. The accuracy and reproducibility of the method has been verified using tubes with various internal diameters in a water tank and repetitive measurements from the common caroid artery of young presumed healthy subjects. The results show that the proposed method is consistent with a standard deviation of 150 micrometers which is on the order of the axial resolution of the ultrasound system used.

Key words: artery diameter, distension, envelope, M-mode, rf-processing, ultrasound.

1. Introduction

Atherosclerosis changes the structure and composition of vessel walls. At a progressed state it may lead to a reduced lumen area (stenosis) and eventually to occlusion. The presence of a stenosis with a diameter reduction of more than 50% can be diagnosed reliably and non-invasively using conventional ultrasound Doppler techniques. These techniques are based on the evaluation of spectral broadening of the Doppler signal originating from the disturbed flow pattern distal to the stenosis. The detection of changes in wall structure and composition without substantial reduction in lumen area requires more refined techniques. Some of these methods are based on the notion that the local elasticity of the vessel wall may be modified by atherosclerotic changes. Moreover, the assessment of the elasticity may provide evidence about the response of the vessel wall to pharmaceutical agents administered, for example, for treatment of hypertension. A third application of elasticity assessment may be the evaluation of the time-dependent wall tonus under the influence of neural, hormonal, and hemodynamic (wall shear stress) activation.

The compliance and distensibility of an artery have been proposed as measures for the elastic behavior of a vessel wall. The compliance C is defined as:

$$C = \frac{\partial V}{\partial p} \tag{1.1}$$

whereas the distensibility D is given by:

$$D = \frac{\partial V}{V \partial p} \tag{1.2}$$

with V the volume of an artery segment and p the blood pressure. Under the assumptions that both quantities are independent of blood pressure and that a change in volume is predominantly caused by a change in diameter rather than elongation of the vessel the above definitions may be modified to:

$$\begin{aligned} CC &= \frac{dA}{dp} \\ DC &= \frac{dA}{A_d dp} \end{aligned} \tag{1.3}$$

where A_d is the cross-sectional area at end-diastole and CC and DC are the compliance and distensibility coefficient per unit of length, respectively. Assuming a circular cross-section the CC and DC can be expressed as a function of diameter at end-diastole and the pulsatile change in diameter due to a pulsatile change in blood pressure. The latter quantity may be difficult to assess non-invasively at the site of measurement and is then replaced by a value measured nearby.

The local change in diameter as function of time (distension waveform) as well as the initial diameter can be measured non-invasively with the use of ultrasound techniques. Hokanson used a zero crossing tracking technique applied to the received radio frequency (rf) signal (HOKANSON *et al.* [9]). A basic problem with this approach is that the initial selection of the signal windows is problematic. It requires positioning of windows on moving structures where the distance between the first zero-crossing within the windows at the anterior and posterior walls provide a direct measure for the time-dependent diameter. Moreover, phase interference of signals originating from closely spaced structure interfaces may cause temporary loss of a zero-crossing. In another approach the rf lines acquired in M-mode are temporarily stored and processed afterwards (HOEKS *et al.* [6], HOEKS [7]). The frozen information allows for accurate positioning of the sample windows while the type of processing employed (Doppler processing, rf correlation) reduces the sensitivity for phase interference because the mean phase of the signal within the window is considered rather than the instantaneous phase. A disadvantage of the latter approach is the short time segment that can be evaluated (usually on the order of 5 seconds). This is insufficient to study, for example, the time-dependent changes in wall tonus.

Apparently there is a need for a processing algorithm capable to locate fast and reliably periodically the wall-lumen interface and to position the sample windows for signal processing. The procedure should require minimal user interaction, has to be rather insensitive to the signal level and should not interfere with the assessment of the change in diameter as function of time. The present paper describes a technique to establish a signal threshold, acting on the envelope of the received of the received signal at enddiastole where the rate of change in diameter is minimal, that adjusts dynamically to the signal level of the nearby walls. At the beginning of a measurement the user has to identify the lumen only once. At the start of each cardiac cycle, signaled by a trigger derived from the R-wave of the ECG, the sample windows are forced outward from the point indicated until the local envelope exceeds the local threshold, giving the initial diameter. From there the displacement detection algorithm will take over resulting in displacement waveforms for both the anterior and posterior walls while the difference between both provides the distension waveform. To ensure that always the signal from the same acoustic interface is evaluated, the sample windows track the structures using the displacement signals observed.

2. Envelope detection

The proposed algorithm for the identification of the wall-lumen interface utilizes the envelope of the rf signal. For a good contrast between wall and lumen (distinct change in amplitude level) the ultrasound beam should be aimed perpendicular to the vessel. Localization of a vessel segment of interest using landmarks as a bifurcation and proper positioning of the M-mode line is executed in 2D B-mode. A commercially available ultrasound scanner (ATL Ultramark-5), operating at 5 MHz, is connected to a real-time data-acquisition system (DAS), developed in our institute. It digitizes with a dynamic range of 8 bits the rf signal received in M-mode synchronously with the emission trigger at a sampling frequency of 20 MHz, i.e., four times the assumed carrier frequency of the ultrasound signal. To reduce the time required for data transfer from the DAS to a computer (PC486-DX2/66) the signal transfer is restricted for a preset depth range starting at a preselected delay with respect to emission. Normally the depth range would be on the order of 10 mm for peripheral vessels. For a signal with a 5 MHz carrier frequency, sampled at 20 MHz, a depth range of 10 mm corresponds with 260 samples per line.

If the rf signal has a narrow bandwidth with respect to the carrier frequency and if the signal is sampled at four times the carrier frequency a subsequent pair of sample values may be considered as a sample of a complex signal. This reduces the computation of the envelope to the square root of the sum of squares of the complex components. For wide bandwidth signals and/or an improper ratio of sample frequency and carrier frequency the instantaneus mismatch will result in a modulation of the envelope related to the instantaneous phase of the received signal. The modulation can be effectively removed by smoothing the calculated envelope with a sliding window with a width corresponding to the axial resolution of the ultrasound system. Figure 1 depicts the calculated envelope using a smoothing over 2 periods, i.e., 8 sample points (0.3 mm). The error due to the assumptions (narrow relative signal bandwith, proper sampling frequency) is hardly visible while the computational complexity is minimal, especially if a look-up table is used for the square root of the sum of squares.

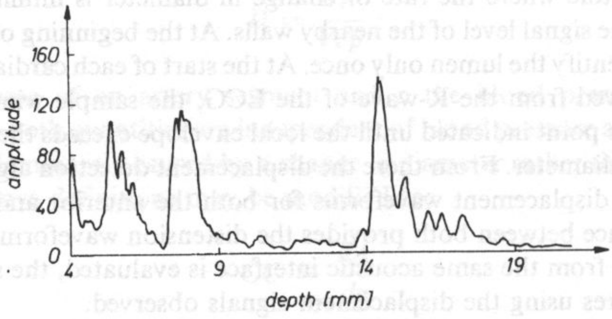

Fig. 1. The envelope of the received signal as function of depth (arbitrary offset, arbitrary amplitude) calculated as the square root of the sum of squares of a subsequent sample pair followed by a smoothing over 8 points (rectangular sliding window).

3. Vessel wall detection

Blood particles have a size considerably smaller than the wavelenght of the ultrasound in the medium and will, therefore, scatter the impinging signal in all directions (point reflector). Moreover, the impedance mismatch of blood particles and the surrounding plasma is low causing only marginal scattering. For both reasons the echogenicity of blood is low. On the other hand the layers of the artery wall of act as strong reflectors. Especially the adventitia returns a strong signal while the signal originating from the intima is considerably lower in amplitude. The media appears usually with a low intensity on a B-mode image of a high resolution system indicating the weak echogenicity of the layer. In Fig. 1 the signal from the adventitia of the anterior and posterior walls of the carotid artery of a young healthy volunteer is clearly visible. Apparently the axial resolution of the system used is too low to show the media and intima as distinct layers. The lumen has a relatively low signal amplitude but close to the anterior wall some reverberated signals can be seen. These artifacts make it difficult to identify the position of the anterior wall-lumen interface.

The echo level of the anterior and posterior walls is different due to a difference in echogenicity or a different gain setting (correction for the depth dependent attenuation). Because the absolute signal level may vary from person to person and from measurement to measurement an absolute signal threshold will be inadequate to detect the position of arterial wall signals. For correct detection the amplitude threshold should dynamically adjust to the value of a local maximum, independent of the local gain setting and intensitive to incidental reverberations or probe movements. This problem may be solved by considering the maximum within a window, with a given width, sliding over the envelope. Then starting from a point somewhere within the lumen, the first crossing in either direction of the actual envelope and a threshold, expressed as a percentage of the local maximum, will indicate the location of the wall-lumen interface. However, the selection of the window width for the determination of the local maximum needs careful consideration. If it is narrower than the lumen diameter, the local maximum may be temporarily dominated by a reverberation. On the other hand, a window width considerably larger than the artery diameter will give preference to either the anterior wall amplitude, whichever is larger. Moreover, a large number of time consuming comparisons will be involved using the sliding window approach discussed above.

Another possibility is to generate a reference signal decaying exponentially as function of depth. If the reference level falls below the current envelope it will be reset to the instataneous amplitude (sustain-attack low pass filter). In this way the reference level is always related to the last dominant echo and it will follow gradual changes in echo level along the ultrasound beam. The decay time (or response time) should be selected according to the expected maximum rate of change in gain level. For most systems this is on the order of 40—60 wavelengths, i.e., for a 5 MHz system about 10 mm. Thus only steep changes in amplitude level, considerably faster than can be explained by changes in gain setting, will be explored for wall-lumen interface detection. Short echoes of low amplitude within the lumen, possibly due to reverberations, will be ignored because they will not reach the threshold level T given as:

$$\begin{aligned} T_i &= \alpha R_i \\ R_i &= (1—2000\frac{f_s}{\beta c})\, R_{i-1} \end{aligned} \tag{3.4}$$

where c is the velocity of sound in the medium (1540 m/s)—, f_s is the sampling frequency given in MHz, i is the sampling index, β is the decay given in mm, and α the fraction (fractional threshold) to convert the reference level R to a threshold T. The parameters α and β may depend on the characteristics of the echo system used. The recursive operation for R is only executed if R_i is greater than the local envelope level (sustain), otherwise it will be set to that level (attack).

The algorithm above will be used to detect adaptively the location of the anterior wall-lumen interface. Starting from somewhere within the lumen (Fig. 2, top) the first

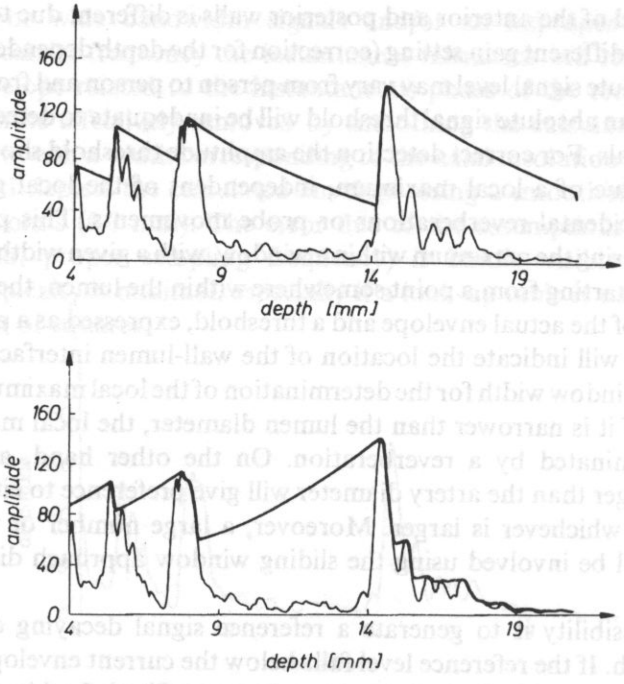

Fig. 2. The reference level (bold) as function of depth in the (top) forward and (bottom) reverse direction with a decay rate of 7 mm, superimposed on the envelope distribution. Starting from within the lumen the first crossing of the envelope and the threshold level (fraction of reference level) indicates the position of the wall-lumen interface.

crossing of the envelope and the threshold level T_i derived from the reference level R_i will signal the position of the wall. This position ought to be corrected for the resolution of the echo system because the trailing edge of the echo is considered. The algorithm will not work for the posterior wall-lumen interface because the threshold level to be applied should be independent of the anterior wall amplitude. To solve this problem a threshold level is generated in the opposite direction (Fig. 2, bottom) and the same procedure as for the anterior wall is followed. No correction is necessary for the axial resolution of the system. The estimated diameter of the blood vessel is then the distance between the detected threshold crossings corrected for the resolution of the system. Subsequently, the observation windows are positioned to track the wall positions and to estimate the wall displacement as function of time. The procedure is repeated once every cardiac cycle, triggered by the R-top of the ECG. This initialization reduces accumulating tracking errors and provides an updated estimate for the diameter on a beat-to-beat base. Of course, user-interaction (identification of a point within the lumen) is only required at the start of a measurement. For the following heart beats the system will be able to select a start point based on the initial setting and the detected (accumulated) changes in wall positions.

4. Evaluation

To validate the proposed processing scheme silicon/rubber tubes with know internal diameters of 8.55, 5.70 and 3.87 mm and with a wall thickness of 1.6, 1.0 and 1.0 mm, respectively, were placed in a water tank at a distance of 2 cm from the transducer face. Measurements were carried out with a 5 MHz ATL Ultramark-5, the rf signal of which was digitized at 20 MHz. The length of the smoothing window for the envelope calculation was set at 4 sample points (1 period). The fractional threshold α was varied from 0.3 to 0.8, while the decay β was varied from 3.8 to 10.6 mm. The experimental results are depicted in Fig. 3, where the estimated diameters are plotted as function of α for different β values without correction for the system axial resolution. Each estimate given is based on the median of 5 independent measurements. It can be concluded from Fig. 3 that for a given α, a smaller decay β results in a smaller estimated diameter. The same holds for a fixed decay: a smaller fractional threshold α gives a smaller estimated diameter. Figure 3 clearly shows that the diameter estimation procedure is more sensitive to the selected threshold α than to the decay parameter β. Considering the bias in the estimate, after correction for an assumed system resolution of 0.3 mm, it can be inferred that a reasonable choice for α would be in the range of 0.5 to 0.7, while the value of the decay parameter should be on the order of the diameter of the tube investigated.

The results, as presented in Fig. 3, indicate that the bias in the estimate after correction for the resolution will be smaller than the resolution of the echo system employed. However, these results can not be used to draw conclusions about the consistency of the estimation algorithm under realistic conditions. For that purpose rf signals from the common carotid artery, about 2 cm proximal to the flow divider, of 4 young healthy volunteers were recorded and used for the evaluation of the algorithm. Eleven arbitrary files were obtained, each of them covering a time period of about 5 seconds (5 to 7 cardiac cycles). Figure 4 depicts the estimated internal diameter, based on the end-diastolic diameter of the second heart beat of file F03, as function of the fractional threshold and decay parameter. The detected diameters are consistent for α values ranging from 0.5 to 0.7 and β values ranging from 6 to 9 mm. These results are in accordance to the results obtained in the tube experiment.

In a further evaluation the parameters were set at $\alpha=0.5$ and $\beta=7$ mm and applied to all beats of all recorded files (estimation of end-diastolic diameter). The results, as listed in Table 1, indicate that the standard deviation of the estimate is on the order of 0.1 mm, implying a good repeatability and stability of the processing algorithm. However, these results can not be used to deduce a possible bias in the estimate. Moreover the observed standard deviation may also originate from true changes in end-diastolic diameter, due to a change in blood pressure level or vessel wall tonus, rather than from an error in the estimation procedure. To eliminate the

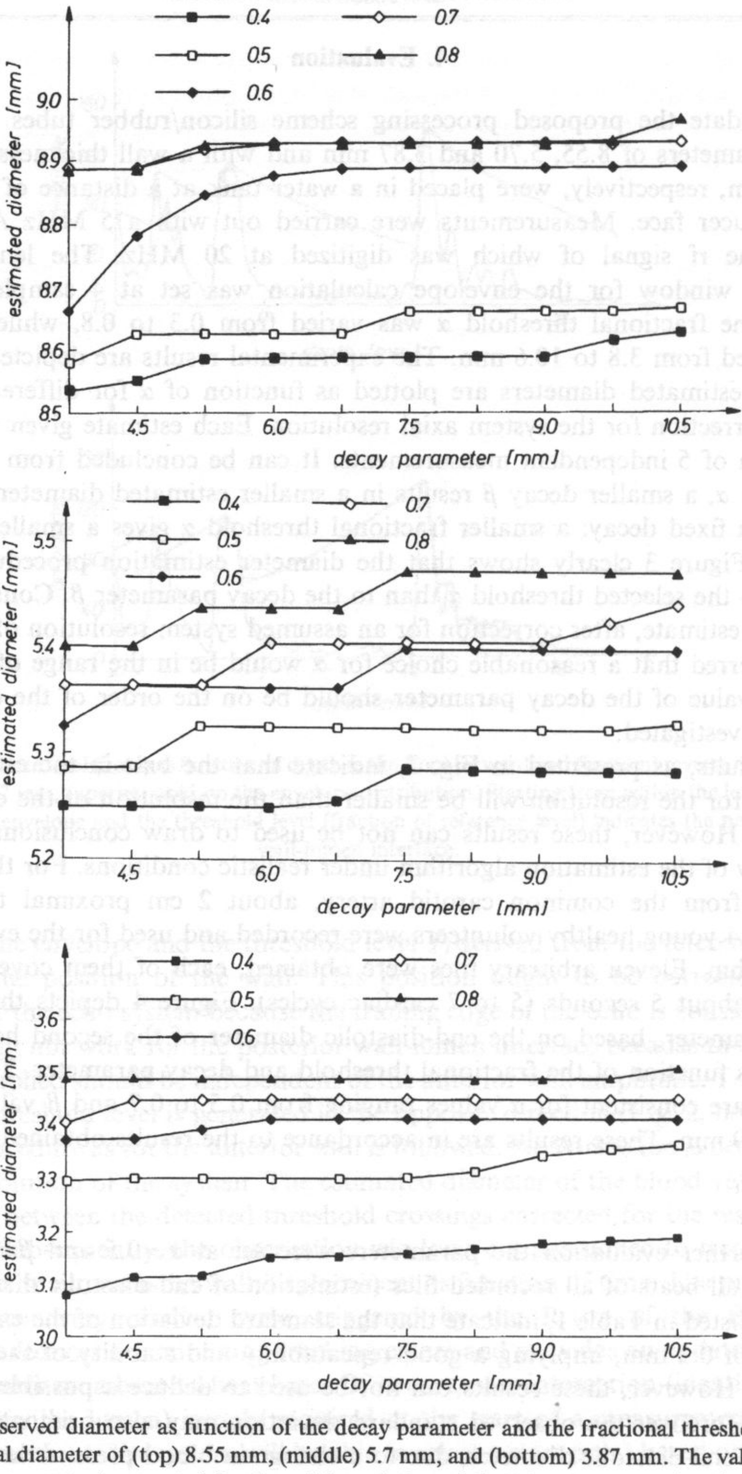

Fig. 3. The observed diameter as function of the decay parameter and the fractional threshold for a tube with an internal diameter of (top) 8.55 mm, (middle) 5.7 mm, and (bottom) 3.87 mm. The values presented are not yet corrected for the axial resolution of the echo system employed.

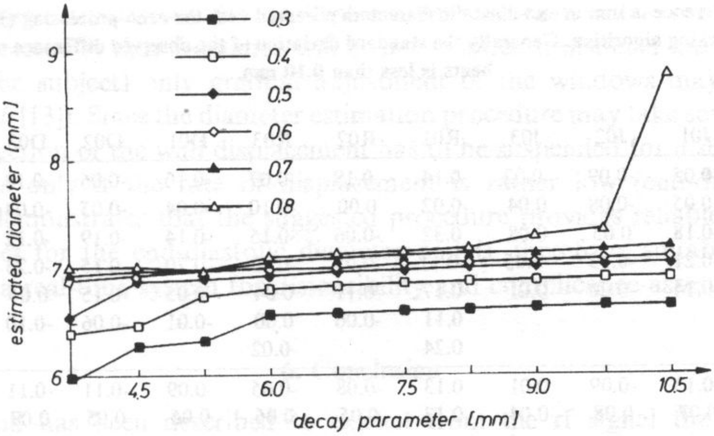

Fig. 4. The estimated diameter of the common carotid artery of a young healthy volunteer at the end of the second heart beat (file F03) as function of the decay parameter and the fractional threshold without correction for the axial resolution of the echo system.

Table 1. The estimated diameter in mm of the common carotid artery of 4 young healthy volunteers for a fractonal threshold of 0.5 mm and a decay parameter of 7.7 mm. Generally, the observed standard deviation is less than 0.15 mm. Note the minor differences for repeated measurements on the same subjects (same letter).

File	J01	J02	J03	R01	R02	R03	D01	D02	D03	F02	F03
beat	5.70	5.31	5.42	6.49	6.92	6.47	6.62	7.11	6.96	6.75	7.17
2	5.66	5.43	5.43	6.68	6.81	6.49	6.60	7.06	6.77	6.85	6.94
3	5.85	5.33	5.44	6.43	6.96	6.43	6.81	7.05	6.98	6.79	7.13
4	5.89	5.54	5.67	6.92	6.97	6.34	6.94	7.09	7.05	6.80	7.05
5	5.81	5.66	5.69	6.76	6.83	6.30	6.85	7.10	6.96	6.75	7.12
6				6.97	6.84	6.34	6.95	7.13	7.02		
7				6.99		6.45					
mean	5.78	5.45	5.53	6.75	6.89	6.40	6.79	7.09	6.95	6.79	7.08
std	0.09	0.13	0.12	0.21	0.06	0.07	0.14	0.03	0.09	0.04	0.08
max-min	0.23	0.35	0.28	0.57	0.16	0.19	0.35	0.08	0.28	0.10	0.23

latter source of error the algorithm for the detection of the wall displacement, based on cross-correlation of rf signal segments (De Jong *et al.* [3] Hoeks *et al.* [7]) was applied. Using this algorithm an estimate for the end-diastolic diameter can be derived from the sum of the initial diameter and the cumulative change in diameter over the cardiac cycle. This value is then compared to the end-diastolic value obtained with the diameter estimation algorithm. The results, listed in Table 2, confirm that the observed difference between both methods and its standard deviation are generally less than 0.1 mm, demonstrating a good repeatability and consistency of both methods. Moreover, comparing Table 2 with Table 1 reveals that the standard deviation of the estimate corrected for a possible change in end-diastolic diameter (Table 2), is indeed systematically lower than the uncorrected one (Table 1),

Table 2. The difference in mm in end-diastolic diameters observed with the echo processing (Table I) and the displacement tracking algorithm. Generally the standard deviation of the observed difference over consecutive beats is less than 0.10 mm.

File	J01	J02	J03	R01	R02	R03	D01	D02	D03	F02	F03
beat 1	-0.05	-0.09	-0.02	0.14	-0.18	-0.07	-0.10	-0.06	-0.18	-0.05	-0.38
2	-0.05	-0.08	0.04	0.02	0.00	-0.10	-0.04	-0.05	-0.04	-0.05	-0.13
3	-0.18	0.05	0.08	0.32	-0.06	-0.15	-0.14	-0.19	-0.12	0.00	-0.20
4	-0.21	-0.13	-0.05	-0.11	-0.05	-0.03	-0.20	-0.15	-0.23	0.01	0.03
5	-0.14	-0.20	0.01	0.17	-0.11	0.04	-0.05	-0.15	0.04	0.11	0.01
6				0.11	-0.06	0.00	-0.01	-0.06	-0.10		
7				0.24		-0.02					
mean	-0.13	-0.09	0.01	0.13	-0.08	-0.05	-0.09	-0.11	-0.11	0.00	-0.13
std	0.07	0.08	0.04	0.13	0.05	0.06	0.06	0.05	0.09	0.06	0.15

confirming that part of the observed random error is caused by true changes in end-diastolic diameter.

5. Discussion

Since the first reported non-invasive measurement of human artery pulsations with ultrasound (ARNDT [1]) many non-invasive studies concerning the mechanical properties of human arteries have been carried out (RENEMAN *et al.* [11]; VAN MERODE *et al.* [12]; BUNTIN and SILVER [2]; LÄNNE *et al.* [10]). The initial artery diameter was determined either by letting the user identify the wall positions using a displayed signal (HOKANSON *et al.* [9]; ERIKSEN [4]; HOEKS *et al.* [6]) or by using a fixed signal threshold to find the positions automatically (ARNDT [1] GUSTAFSSON *et al.* [5]; WILSON *et al.* [13]). Both methods have obvious drawbacks. Unless the signal is frozen, accurate manual identification of vessel wall positions is problematic, while a fixed threshold will fail if the vessel wall signal is not distinct, is too low or varies considerably over time.

The proposed detection algorithm intends to improve the diameter detection procedure in two aspects. Firstly, to provide a method that can detect the wall positions automatically on a beat-to-beat base, allowing for a continuous recording of the change in diameter as function of time. Secondly, a dynamic threshold signal is employed that adjusts its level to the level of the current signals. The only user interaction required is a rough identification of the position of the lumen at the start of a measurement. In a practical configuration a computer screen would display the envelope of the captured rf signal repetitively allowing the user to adjust the signal level and to identify the lumen. The last action starts the estimation procedure for both the internal diameter as well as the change in wall positions over time yielding the distension waveform. The diameter estimation procedure will be carried out at the start of each cardiac cycle signaled by a trigger derived from the R-wave of the ECG.

Thus every cycle the windows for the displacement algorithm will be repositioned. To reduce the possibility that the lumen will be lost (sudden transducer movements with respect to the subject) only gradual adjustment of the windows may be allowed (WILSON *et al.* [13]). Since the diameter estimation procedure may take some computer time the detection of the wall displacement has to be suspended for a short time but this happens only if the rate of displacement is rather low (end-diastole). The evaluation demonstrates that the suggested procedure provides reliable and consistent estimates for the end-diastolic diameter and is, therefore, suitable for incorporation in a real-time system for distensibility and complicance assessment.

6. Conclusion

A method has been described to extract from the rf signal the diameter of a peripheral vessel at end-diastole. The threshold level adjusts dynamically to the signal level of nearby vessel walls, making the procedure insensitive to the effects of gain setting and depth dependent attenuation. In *in vitro* and *in vivo* test measurements the proposed procedure showed a good consistency of 0.15 mm, which is on the order of the axial resolution of the ultrasound echo system used. The selection of the processing parameters (decay and fractional threshold) is not critical. The computational complexity is only moderate allowing for the incorporation of the estimation algorithm in real-time distensibility and compliance measurement systems.

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