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Automatic Measurement of Lumen Diameter of Carotid Artery in A-Mode Ultrasound

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Abstract— **Accurate measurement of lumen diameter is essential for correct estimation of arterial compliance. We have been developing a new non-invasive arterial compliance measurement tool using a single element ultrasound transceiver. In this paper we propose a new method for measurement of lumen diameter from single line of Radio-Frequency Signal (RF) obtained from the common carotid artery (CCA). The method is free from fixed thresholds and uses shape fitting to get objective measurement. The accuracy of the algorithm was found to be better than 5 % for software simulated and phantom arteries and better than 10 % in case of data obtained from CCA of human volunteers.**

I. INTRODUCTION

Stiffness of CCA is well known indicator of cardiovascular disease [1-2]. Equation (1) gives the compliance coefficient (CC) of artery [3].

$$\mathrm{CC} = \frac{\Delta \mathrm{D} \times \mathrm{d_d}}{2 * \Delta \mathrm{P}} \quad (1)$$

$\mathrm{d_s} =$ Systolic lumen diameter
$\mathrm{d_d} =$ Diastolic lumen diameter
$\Delta\mathrm{D} = \mathrm{d_s} - \mathrm{d_d}$
$\Delta\mathrm{P} =$ Systolic pressure $-$ Diastolic pressure

It can be clearly seen from (1) that, measurement of lumen diameter of the artery is an important step in calculating arterial compliance. The first step of compliance measurement is identification of Region of Interest (ROI) either manually [4], or automatically [5]. We are in the process of developing a portable arterial compliance measurement tool employing a single element ultrasound transducer [6]. We call this system ARTSENS. We had in the past demonstrated a method for automatic identification of approximate location of proximal wall (PW) and distal wall (DW) of the CCA [5] in ARTSENS. In this paper we propose a method for measurement of lumen diameter of the artery accurately and objectively when approximate locations of PW and DW are known. CCA is a layered structure with three layers - adventia, media and intima – generally having different levels of echogenicity [3]. The lumen diameter is the distance between the trailing edge of intima of PW and leading edge of intima of DW. Measurement of correct diastolic diameter d_d is necessary for accurate compliance measurement. We had earlier used a method in ARTSENS for diastolic diameter [6] that basically involved thresholding the double derivative of points corresponding to the intima.

Very few other works have tried to find the lumen diameter of the artery in past from A-Mode data [7-8]. As far as we have explored almost all other methods reported in literature work, exclusively with B-Mode images. The previous method of diameter estimation in our system sometimes ran into problems because the threshold level for the rate of rise of amplitude for intima cannot be objectively determined. In this paper we propose a new method of diameter estimation by using dynamic peak detection and shape fitting which is less prone to errors and more objective. The performance of the new algorithm was verified by simulation and phantom studies and measurements conducted on a 16 human volunteers.

II. DATA ACQUISITION

The data is acquired using ARTSENS system which consists of a single element ultrasound transceiver operating at 5 MHz [6]. The transceiver is operated in pulse-echo modality. The pulse rate is 50 pulses per second. The received echoes were digitized at a rate of 100 Mega-samples per second (Msps). After each pulse, received signals are digitized for about 52 µs so that the depth of vision is about 40 mm taking the average speed of sound in tissue to be 1540 m/s. Each millimeter is about 130 points (PPMM) and thus each frame consists of about 5200 points. A typical frame from CCA is shown in Fig. 1.

III. DIAMETER MEASUREMENT ALGORITHM

The first step in measurement of compliance is to locate approximate positions of PW and DW which was done using the method discussed in [5]. This step gives us two approximate sample numbers corresponding to proximal wall (pw) and distal wall (dw) on frames containing the CCA.

A. Extraction of Arterial Well

We extract the samples corresponding to the lumen of the artery from any frame (F) and call it as arterial well (AW).

$$\mathrm{AW} = \mathrm{F}\begin{pmatrix}\mathrm{pw} - 1.5 \times \mathrm{PPMM}, \mathrm{pw} - 1.5 \times \mathrm{PPMM} + 1\,, \\ \ldots, \mathrm{dw} + 1.5 \times \mathrm{PPMM}\end{pmatrix} \quad (2)$$

AW is represented by the data enclosed in the dotted rectangle in Fig. 1.

B. Extraction of Arterial Halves and Alignment

The initial estimate for the mid-point (MPI) of AW is

$$\mathrm{MPI} = \frac{\mathrm{dw} + \mathrm{pw} + 3 \times \mathrm{PPMM}}{2} \quad (3)$$

AW is cut into two halves left half (LHI) and right half (RHI) as indicated in (4) and (5).

$$\mathrm{LHI} = \mathrm{AW}(1,2,\ldots,\mathrm{MPI}) \quad (4)$$

$$\mathrm{RHI} = \mathrm{AW}(\mathrm{MPI}+1, \mathrm{MPI}+2, \dots, 2 \times \mathrm{MPI}) \tag{5}$$

Sequence of data points in LHI is reversed to get LHIR. MPI is only the approximate mid-point of AW. PW and DW are of the same artery so they will have similar characteristics. Using this clue we calculate the shift *S* between LHIR and RHI by using cross-correlation. We pad *S* zeros to the beginning of LHIR or RHI appropriately so that the wall in each sequence is nearly at the same distance from the beginning of the sequence (Fig. 2).

Now each element of LHIR and RHI are squared (6) and (7) to get $\mathrm{LH_{sq}}$ and $\mathrm{RH_{sq}}$ which suppresses the lumen noise and enhances the wall features Fig. 2.

$$\mathrm{LH_{sq}} = \mathrm{LHIR}^2 \tag{6}$$

$$\mathrm{RH_{sq}} = \mathrm{RHI}^2 \tag{7}$$

$\mathrm{LH_{sq}}$ and $\mathrm{RH_{sq}}$ are both filtered with a 1 MHz low-pass butter-worth filter of 1st order. This eliminates the high frequency components keeping the variations in signal due to wall boundaries intact. This is followed by a normalization of amplitude between 0 and 1 for each sequence, to get LH and RH respectively as shown in Fig. 3.

B. Dynamic Peak Detection

As CCA is a layered structure there can be multiple echo peaks at each wall corresponding to reflections from different layers. The lumen is characterized by its relatively flat and low-amplitude characteristics. In LH or RH there may be one to three peaks that are distinctly visible and number of visible peaks may differ between frames.

The peak detection is based on the fact that a peak is followed by samples of lower amplitude than itself at-least until the successive distinct valley [9]. If a distinct peak is having amplitude $\mathrm{p_k}$ and is located at $\mathrm{x_p}$ and successive distinct valley having the amplitude $\mathrm{v_k}$ is present at $\mathrm{x_v}$ $(\mathrm{x_v} > \mathrm{x_p})$ then there will always be a sample having amplitude a between x_p and x_v such that $\mathrm{a} - \mathrm{p_k} > \mathrm{T}$ where T is a threshold value. Similarly a distinct valley is followed by samples of higher amplitude until at-least the successive distinct peak and is detected by similar procedure using the same threshold T. T is initially set to a very low value (0.1). Peaks are detected for both LH and RH by setting T as the threshold. CCA is always greater than 4 mm in diameter [10]. If number of peaks detected for LH beyond 2 mm is less than 4 then the nearest peak location is stored as $\mathrm{NP_l}$. If numbers of such peaks are more than 3 then T is increased by 0.01 and the procedure is repeated iteratively until $\mathrm{NP_l}$ is located. The same procedure is followed for the right-half RH to find $\mathrm{NP_r}$ (Fig. 3).

C. Shape Fitting and Diameter Estimation

Now, Lumen of right half (LRH) and lumen of left half (LLH) are extracted from RH and LH (8) (9).

$$\mathrm{LLH} = \mathrm{LH}(1, 2 \dots, \mathrm{NP_l}) \tag{8}$$

$$\mathrm{LRH} = \mathrm{RH}(1, 2, \dots, \mathrm{NP_r}) \tag{9}$$

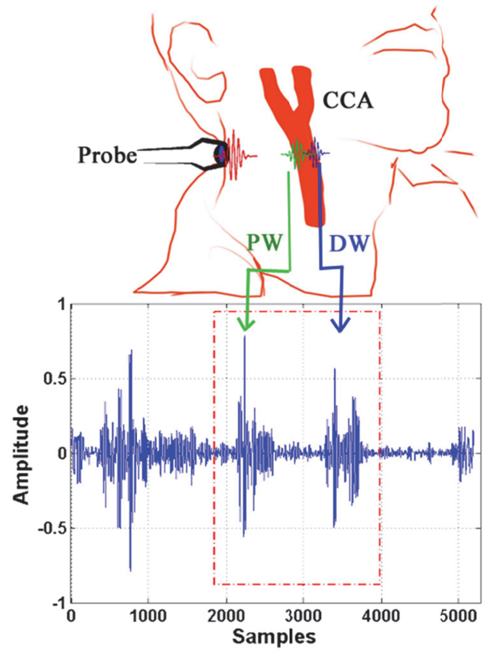

Figure 1. A typical frame obtained from CCA in ARTSENS using a single element ultrasound transciever. The dotted rectangle shows the arterial well (AW) that is culled for diameter measurement.

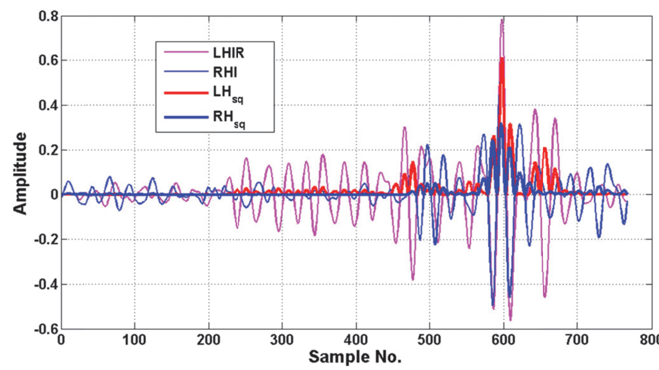

Figure 2. Left half and right half were aligned by padding *S* zeroes to any one half and then square of each sample point was calculated for both curves.

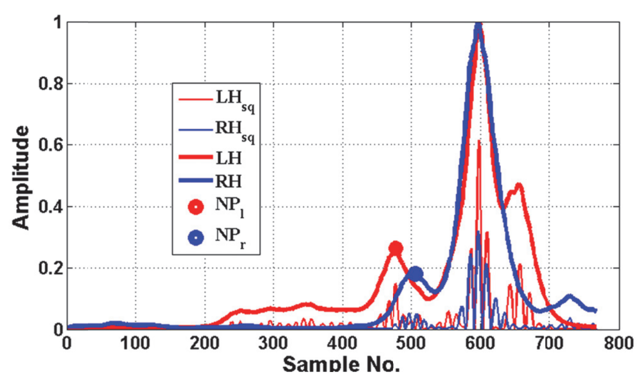

Figure 3. Left half and right half were filtered, normalized and their innermost significant peaks were detected.

Each lumen half has a shape corresponding to a horizontally flipped L with a slanting rising edge. We try to fit a function FLR to LRH. FLR is a flipped L shape defined by 3 points (10).

$$\mathrm{FLR}(1) = \mathrm{LRH}(1)\,,$$
$$\mathrm{FLR}(\mathrm{RP_l}) = \mathrm{LRH}(\mathrm{NP_r}) \qquad (10)$$
$$\text{and } \mathrm{FLR}(\mathrm{n}) = \mathrm{LRH}(\mathrm{n}),\ \ 1 < \mathrm{n} < \mathrm{NP_r}$$

FL is defined by two line segments between $[(1, \mathrm{FLR}(1)), (\mathrm{n}, \mathrm{FLR}(\mathrm{n}))]$ and $[(\mathrm{n}, \mathrm{FLR}(\mathrm{n})), (\mathrm{NP_r}, \mathrm{FLR}(\mathrm{NP_r}))]$. n is varied between 2 to $\mathrm{NP_r} - 1$ so that we have minimum square error between FLR and LRH. This optimal n is the leading edge of the DW (LEDW). The same shape fitting procedure is repeated for LLH by Similarly, another flipped L shape FLL is fit to LLH and trailing edge of PW (TEPW) is identified (Fig. 4). Finally diameter is calculated by (11).

$$\text{Arterial diameter } (D) = \frac{\mathrm{LEDW} + \mathrm{TEPW} - S}{PPMM} \qquad (11)$$

IV. Validation of The Algorithm

The algorithm was validated by verification of accuracy of estimated diameter. Three methods of verification were employed.

i. Measurement on simulated arteries.

ii. Measurement on five phantom arteries.

iii. Measurement on human volunteers and comparison with manual measurements.

A. Simulation Studies

A simulation platform was designed that could generate simulated frames with echoes placed at desired locations. Noise recorded from ARTSENS hardware was amplified and overlaid on the simulated signals to produce signals with desired SNR. Frames were simulated with lumen diameters (D_s) ranging from 5 mm to 10 mm and SNR ranging from -5 dB to 15 dB. Automatic diameter estimation on these frames were accurate by better than 2 % for SNR above 0 dB (Fig 5).

B. Phantom Studies

Phantoms of CCA were constructed by dipping silicone pipes of different inner diameters and wall thickness into transparent plastic cylinders filled with water (Fig. 6).

Five frames each were acquired for five different pipes ($F_1, F_2, \ldots, F_5$). Diameter of each pipe was automatically measured from 5 frames as ($AD_1, AD_2, \ldots, AD_5$). Mean automatic diameter of each pipe was calculated by (12).

$$D_m = \frac{(AD_1 + AD_2 + AD_3 + AD_4 + AD_5)}{5} \qquad (12)$$

Actual inner diameter of each pipe was measured by using vernier-caliper (D_V). It was found that the measurement of diameter was accurate by better than 4 % for all pipes (Table I).

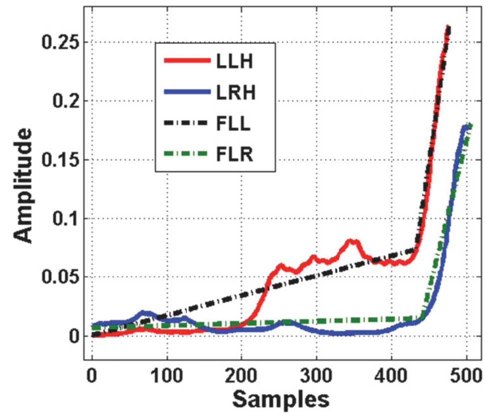

Figure 4. Flipped L shape are fit to the lumen halves. The vertex of the L are declared as the base of the echo from intima.

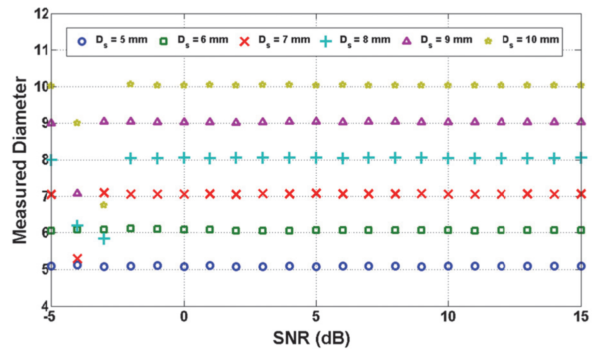

Figure 5. The measured diameters are very close to the simulated diameters with error of less than 2 % at SNR as low as 0 dB.

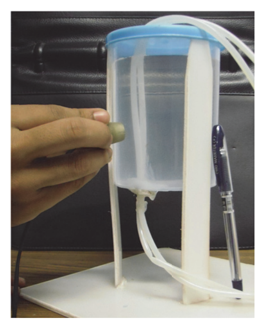

Figure 6. Image shows the arrangement of the two silicone pipes inside water. Method of placing the probe is also shown.

TABLE I. Comparison of Manual and Automatic Measurements of Phantom Pipes.

Pipe No.	1	2	3	4	5
AD_1 (mm)	4.46	5.39	12.37	3.92	6.28
AD_2 (mm)	4.46	5.39	12.37	3.91	6.29
AD_3 (mm)	4.46	5.39	12.39	3.91	6.28
AD_4 (mm)	4.46	5.39	12.37	3.92	6.29
AD_5 (mm)	4.45	5.39	12.37	3.92	6.29
D_m (mm)	4.46	5.39	12.37	3.92	6.29
D_v (mm)	4.34	5.45	12.40	4.00	6.50
Error (%)	3.0	1.2	0.2	3.0	3.2

C. Human Volunteer Study

Diameter was estimated for 16 human volunteers by using the automatic algorithm. Ten frames containing the CCA were analyzed per volunteer. In non-invasive measurement using ultrasound ground truth is unavailable. CCA diameter for each frame was manually measured by an operator and then it was compared with automatic measurements. The average mismatch between manual and automatic diameter was found to be less than 10 % (Table II).

TABLE II. COMPARISON OF MANUAL AND AUTOMATIC MEASUREMENTS OF CCA DIAMETER OF HUMAN VOLUNTEERS.

Volunteer No.	Mean Manual Diameter (mm)	Mean Automatic Diameter (mm)	Mean Error (%)
1	6.04	6.295	4.214
2	5.496	5.765	9.351
3	5.519	5.75	6.844
4	6.469	6.599	9.16
5	5.65	5.95	5.405
6	5.488	5.858	6.746
7	8.252	8.611	4.348
8	5.965	5.695	8.053
9	6.263	6.589	5.216
10	5.51	5.886	6.841
11	5.432	5.494	3.054
12	3.928	3.927	2.54
13	5.464	5.729	4.86
14	6.337	5.896	9.28
15	4.966	5.132	3.369
16	5.913	5.864	1.22

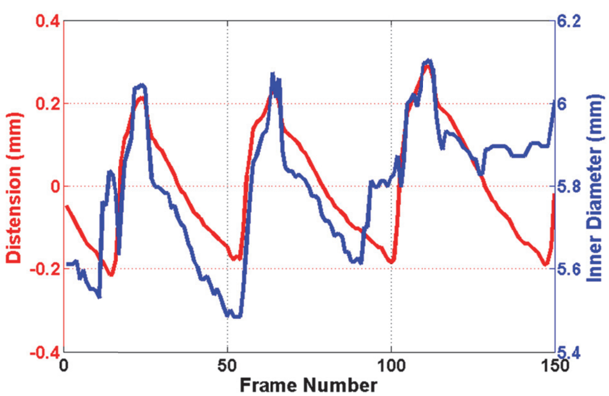

Figure 7. Distention waveform correlates well with the lumen diameter waveform.

Arterial distension was measured by estimating the shift of both walls between successive frames using correlation method [6]. The method uses information from multiple frames to find the accurate distention curve but this cannot give us the lumen diameter. The algorithm of automatic lumen diameter discussed in this paper gives us a noisier curve because the blood-intima interface is generally not well defined. But it was also seen that the shape of the lumen diameter waveform over successive frames correlated well with the distention motion of artery (Fig. 7). This serves as an additional validation for the correctness of lumen diameter.

V. CONCLUSION

A new method for measurement of CCA diameter from single line of RF obtained from CCA was proposed. This method is based on dynamic thresholding and shape fitting and can be used to obtain objective measurements with minimal sensitivity to noise. The accuracy of the method was found to be better than 2% for software simulated arteries and better than 5 % for phantom arteries made with silicone pipes. The algorithm could measure the diameters of CCA of 16 human volunteers automatically and the measurements agreed by better than 10% with the manual readings. As an additional validation it was seen that the shape of lumen diameter waveform correlated well with the shape of the distension curve.

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