ELSEVIER

Ultrasonics 38 (2000) 369–375

Ultrasonics

www.elsevier.nl/locate/ultras

Speckle tracking for multi-dimensional flow estimation

L.N. Bohs [a,*], B.J. Geiman [a], M.E. Anderson [a], S.C. Gebhart [b], G.E. Trahey [a]

[a] Department of Biomedical Engineering, Duke University, 136 Hudson Hall, Durham, NC 27708, USA
[b] Department of Biomedical Engineering, Vanderbilt University, 5824 Science & Engineering Building, Nashville, TN 37235, USA

Abstract

Speckle tracking methods overcome the major limitations of current Doppler methods for flow imaging and quantification: angle dependence and aliasing. In this paper, we review the development of speckle tracking, with particular attention to the advantages and limitations of two-dimensional algorithms that use a single transducer aperture. Ensemble tracking, a recent speckle tracking method based upon parallel receive processing, is described. Experimental results with ensemble tracking indicate the ability to measure laminar flow in a phantom at a beam–vessel angle of 60°, which had not been possible with previous 2D speckle tracking methods. Finally, important areas for future research in speckle tracking are briefly summarized.

Keywords: Angle-independent flow; Flow quantification; Flow velocity; Vector velocity

1. Introduction

Doppler ultrasound has become widespread for imaging flow and quantifying blood and tissue velocities. However, its uses are limited because it only quantifies the axial component of motion, i.e. that along the direction of the transmitted acoustic wave. In addition, aliasing limits the maximum velocity range and can lead to incorrect velocity assessment [1].

Speckle tracking methods overcome these limitations by directly tracking the backscattered echoes produced by ultrasonic scatterers in blood and tissue. Such speckle patterns, or 'signatures', remain relatively constant as the blood or tissue moves, and can thereby be tracked using pattern matching techniques. This pattern matching can be performed in one, two, or three dimensions, depending on the dimensions of ultrasound data available. Of course, one-dimensional methods do not overcome the angle dependence of Doppler, but they provide some other advantages [2].

Fig. 1 shows the geometry of the pattern matching search used in 2D speckle tracking. A kernel region is identified in a first acquisition, and tracked within a surrounding search region in a later acquisition. The size of the kernel defines the spatial velocity resolution, while the size of the search region relative to the kernel defines the velocity range. The best match between the kernel and the search is determined using a pattern matching search among possible matching locations. The location of the best match defines the vector of motion for that particular kernel. Given the time between acquisition of the kernel and search regions, the 2D velocity vector is calculated. When this is done over many small kernels throughout a region of interest, a vector velocity map is generated. Updating this procedure continuously through time allows examination of temporal velocity variations, similar to those provided by pulsed-wave (PW) or color flow (CF) Doppler, but without angle dependence. A similar procedure is followed for 3D speckle tracking, except that the kernel and search regions are formed from volumetric data acquired from a 2D transducer array [3].

One of the early objections to speckle tracking methods was their computational complexity. However, efficient algorithms and advances in processing power have made multidimensional speckle tracking methods within the realm of real-time operation [4].

In the next section, we outline the development of speckle tracking methods, with a focus on 2D velocity estimation and the contributions of our group at Duke University. In addition, while methods employing multiple transducers have been proposed [5], we limit

* Corresponding author. Tel: +1-919-660-5155; fax: +1-919-684-4488.
E-mail address: larry.bohs@duke.edu (L.N. Bohs)

0041-624X/00/$ - see front matter
PII: S0041-624X(99)00182-1

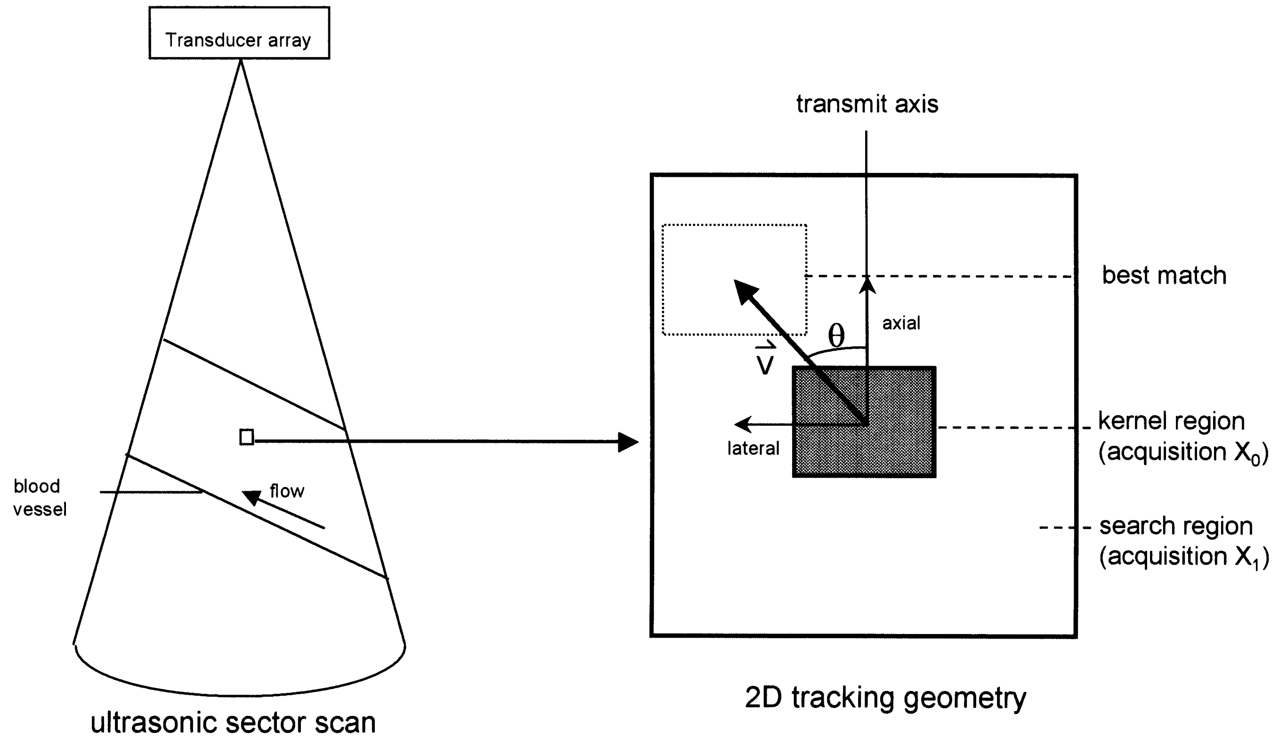

Fig. 1. 2D speckle tracking geometry. A motion field is examined in two separate acquisitions. In the first, a small kernel region is identified. This kernel is tracked within a larger search region in a later acquisition. The best match to the kernel within the search defines the vector of motion for that acquisition interval. Typically, the kernel and search regions are small relative to the size of the motion field in order to provide fine spatial velocity resolution. Note also that the kernel and search regions are not necessarily square, but may take on any rectangular dimensions to provide the desired spatial resolution and velocity range. Both the axial and lateral components of the motion vector are defined, in contrast to Doppler methods, which define only the axial component.

this discussion to methods that use a single transducer aperture, such as that in standard phased array probes.

2. A brief history of speckle tracking

The possibility that speckle patterns from moving objects could be tracked to measure their 2D motion was first reported by Robinson et al. [6], and later by Akiyama et al. [7] and our group [8].

Our report of this phenomenon as applied to blood flow measurement involved digitized images acquired from the popliteal vein, using a 10 MHz transducer [8]. Kernels with a physical size of 2 mm × 1.5 mm were tracked using the normalized correlation algorithm

$$\rho(\alpha,\beta)=\frac{\sum_{i=1}^{l}\sum_{j=1}^{k}[X_0(i,j)-\bar{X}_0][X_1(i+\alpha,j+\beta)-\bar{X}_1]}{\sqrt{\sum_{i=1}^{l}\sum_{j=1}^{k}[X_0(i,j)-\bar{X}_0]^2\sum_{i=1}^{l}\sum_{j=1}^{k}[X_1(i+\alpha,j+\beta)-\bar{X}_1]^2}}, \quad (1)$$

where X_0 and X_1 contained the kernel and search regions respectively, $l\times k$ were the (lateral × axial) dimensions of the kernel region, and (α, β) were the coordinates of a trial matching region in X_1. $\bar{X}_0$ and $\bar{X}_1$ are the spatial mean pixel values of the corresponding image regions. This qualitative study demonstrated that the general direction of 2D motion could be discerned, and that small vectors could designate the magnitude and direction of the measured blood flow velocities.

Later experiments investigated performance variations with respect to the size of the kernel regions utilized. Results showed that performance improved as the kernel was made larger, since the kernel was thereby more unique and less susceptible to noise [9,10]. Of course, increasing the size of the kernel degrades spatial resolution, leading to one of the fundamental tradeoffs in speckle tracking of bulk motion: as the kernel size increases, accuracy and precision generally improve but spatial velocity resolution degrades. As we will see later, however, the kernel size versus accuracy relation does not necessarily hold when speckle tracking is used to measure velocities in the presence of flow gradients.

Another fundamental parameter critically affecting speckle tracking performance is the type of ultrasonic echo data utilized: radio frequency (RF) or detected. Initial experiments with a speckle-generating sponge phantom indicated that RF data provided improved performance [9]. However, later experiments with a flow phantom provided conflicting results, as described below.

Our desire to build a real-time 2D speckle tracking system led us to investigate pattern matching algorithms that were more computationally efficient than normalized correlation. One such algorithm, sum-absolute-difference (SAD) [11], was found to perform similarly to normalized correlation in a quantitative study utilizing known translations of a speckle generating phantom, as well as in qualitative comparisons of velocity maps obtained from flowing blood and moving tissue [12]. The SAD algorithm is far simpler to implement in digital hardware than normalized correlation, requiring only a single absolute difference operation:

$$\epsilon(\alpha, \beta)=\sum_{i=1}^{l} \sum_{j=1}^{k} |X_0(i,j)-X_1(i+\alpha, j+\beta)|. \tag{2}$$

The quantity ϵ is referred to as the SAD coefficient, representing the 'error' in matching a kernel region in acquisition X_0 with a trial matching region in a search region in acquisition X_1. $l \times k$ are the (lateral $\times$ axial) kernel region dimensions in pixels, and (α, β) are the (lateral, axial) coordinates of the trial matching region relative to the center of the search region. The (α, β) producing the minimum ϵ are used to compute the velocity magnitude and angle, given the time between acquisition of the kernel and search regions.

Given the positive results with the SAD algorithm, we constructed a system to compute 2D velocity maps continuously and in real time [4]. This system was based on up to 15 SAD tracking boards operating in parallel, each of which processed digital samples through the SAD algorithm at 25 MHz, the axial sampling rate. Experiments with a string phantom and a $1.0\ \text{mm} \times 2.0\ \text{mm}$ kernel demonstrated accurate quantification of velocities up to $2.5\ \text{m s}^{-1}$ in any direction in the scan plane. A qualitative in vitro flow experiment, using a $0.8\ \text{mm} \times 0.8\ \text{mm}$ kernel, demonstrated the ability to display lateral flow with a 2D color map.

To maximize temporal resolution and velocity range, a custom sequence of B-mode lines was utilized, rather than a pair of full B-mode scans. This custom sequence divided the scan sector into a number of vector lines, each comprising one vector laterally and the entire region of interest axially. Each vector line thus required the acquisition of only $l+m$ beam lines, where l was the lateral pixel size of the kernel and m was the lateral size of the search region. For example, if the kernel size was 2×10 and the search region size was 8×30 pixels, then the custom sequence would involve acquiring the two kernel beams, then the eight search beams. The kernel was typically centered spatially with respect to the search region, but could be offset to one side to provide a larger velocity range in one lateral direction, similar to the baseline shift sometimes employed in Doppler velocity measurements. Similarly, a variable delay was included between the acquisition of the kernel and search region beams to scale the velocity range to the expected target velocities.

Velocity calculations in the real-time system were complicated because kernel and search regions were acquired one line at a time, as in a typical B-mode scan. This required the system to determine the time interval between acquisition of the last kernel beam and acquisition of the last beam of its best match in the search region. In addition, higher velocities were detectable for targets moving in lateral opposition to the beam acquisition direction; this resulted in an asymmetric range of measurable velocities. Nevertheless, the real-time system was effective at quantifying and displaying 2D velocities, particularly of bulk motion such as string and sponge phantoms, and of moving tissue.

In experiments with a laminar flow phantom, the real-time speckle tracking system yielded variable results [13]. For pure lateral flow (i.e. a beam–vessel angle of 90°), velocity and volumetric flow measurements were accurate and consistent over a wide range of flow rates when detected data was used. Conversely, large performance variations were observed for small variations in the transducer-to-flow angle when RF data was used. The inconsistency in performance with RF data was caused by an insufficient axial sampling rate relative to the transmitted frequency, and the subsequent quantization of the tracking grid. This phenomenon occurred even when the axial sampling rate was about twice the Nyquist limit. Nevertheless, RF data can provide better performance than detected data if the axial sampling rate is sufficiently high. Conversely, using RF data increases the likelihood of false peak errors [9], which may arise because of the cyclic nature of the RF signal.

The main limitation of the real-time system was its inability to quantify velocities accurately when the flow had a significant axial component [13]. Velocity profiles measured in a laminar flow phantom rapidly degraded as the flow angle changed from 90° to 105° to 120°. This anomaly was understood in terms of the vast mismatch in spatial frequencies between the lateral and axial dimensions in ultrasonic speckle. The axial spatial frequencies are centered at the relatively high transmit frequency, while the lateral spatial frequencies are centered at DC. This fundamental difference in spatial frequency content makes speckle patterns far more susceptible to decorrelation as a result of relative scatterer motion in the axial direction. This characteristic has strong implications for speckle tracking in flow measurement: flow velocities are increasingly more difficult to measure as the beam–vessel angle approaches the transducer axis [14]. Such relative scatterer motion is more prevalent as the velocity gradient increases; for example, the gradient is highest near the vessel walls in laminar flow. Even for plug flow, this phenomenon makes measurement of velocities prone to errors near the vessel walls.

In addition, increasing the size of the kernel does not necessarily improve accuracy in flow measurement, as with bulk motion. By contrast, increasing kernel size may degrade accuracy, especially in regions of large flow gradients, since the range of velocities present in the kernel is greater, which increases the rate of speckle pattern decorrelation [15].

The realization that flow gradients were the major source of speckle decorrelation led to the development of algorithms that would minimize decorrelation by tracking over smaller times and distances. The only way to do so with line-by-line acquisition was to decrease the lateral size of the kernel and search regions. Even so, multiple acquisitions would be required, and the problems of asymmetrical velocity range would still be present. Performance with line-by-line acquisition was further degraded by speckle motion occurring as the multiple beams for each vector line were acquired. Such speckle motion effectively distorted the speckle patterns, with the degree and shape of the distortion dependent upon the motion direction relative to the scanning direction.

We hypothesized that parallel receive processing [16] would solve these problems by allowing multiple receive beams to be acquired simultaneously. In this way, the kernel and search regions could be acquired in two 'snapshots'. So doing would minimize the time between acquisitions, as well as provide symmetrical lateral velocity ranges and avoid scanning distortion.

For the past few years we have been investigating a form of speckle tracking called ensemble tracking [17]. Ensemble tracking involves multiple parallel acquisitions along a line of sight. Between each acquisition pair, a kernel in the first snapshot is tracked within a search region in the second. The resulting grid of SAD coefficients is added to those resulting from tracking between all other acquisition pairs, and this summed grid is used to determine the speckle pattern translation. Such coherent averaging dramatically improves performance by reducing the effects of random noise. Ensemble tracking is similar in concept to a method proposed earlier by Bonnefous for 3D velocity measurement [18]. Bonnefous's method used 1D correlations between two pairs of three beams arranged in orthogonal planes, with one beam common to the center.

Given that commercial ultrasound scanners currently employ at most 4:1 parallel receive processing, lateral interpolation was necessary to provide sufficient velocity quantization. For example, using a two-line kernel within a four-line search region provides possible lateral translations of only -1, 0, or $+1$ pixel, which are sufficient to determine lateral direction but not lateral magnitude. Commonly used interpolation algorithms were investigated but found to perform poorly, especially when the minimum of the SAD grid was in one of the outer columns, meaning that the actual translation magnitude was greater than 0.5 pixels. The grid slopes interpolation algorithm [19] was developed to address this unique problem.

Grid slopes uses SAD coefficients from two grid locations to interpolate translations to sub-pixel accuracy. Conceptually, Grid slopes relies on the following assumptions, assuming for the sake of explanation a purely lateral translation. (1) As a speckle pattern translates between zero and one pixel, the two SAD coefficients bounding the actual translation vary linearly between a minimum of zero and a maximum of ϵ_{max}, while the sum of the two SAD coefficients remains constant at ϵ_{max}. For example, one SAD coefficient increases from 0 to ϵ_{max}, the other decreases from ϵ_{max} to 0, and both are equal to $\epsilon_{max}/2$ when the translation is 0.5 pixels. (2) ϵ_{max} can be approximated by a SAD coefficient on the zero-lag grid. This grid is computed by tracking a kernel within a search region from the same ultrasonic acquisition. The zero-lag SAD coefficient provides a measure of the energy in the local region, which provides a scaling factor for the algorithm.

In experiments with a sponge phantom, ensemble tracking with the grid slopes algorithm provided excellent performance at transducer angles of 0, 45 and 90° and over three parallel receive beam spacings. Initial in vivo results demonstrated promise for tracking lateral flow in the carotid artery [17]. Both of these studies utilized kernel regions as small as $0.1\ \mathrm{mm} \times 0.2\ \mathrm{mm}$. More recently, ensemble tracking using only 2:1 parallel receive processing was found to perform better than that using 4:1 parallel receive, particularly at small target translations [20]. The difference in performance was attributed to the improved symmetry between beams in 2:1 parallel receive.

Most importantly, recent results evaluating ensemble tracking for laminar flow measurement indicate that it provides far better performance than older speckle tracking methods in the presence of velocity gradients, particularly when the flow has an axial component [21]. For example, Fig. 2 shows a velocity profile measured at an angle of 60° in a laminar flow model with a flow rate of $8\ \mathrm{ml\ s^{-1}}$. The kernel size was 1×20 pixels, and the search size was 2×40 pixels. Volumetric flow of $7.6 \pm 0.5\ \mathrm{ml\ s^{-1}}$ was computed from this velocity profile by integration, assuming circular symmetry. Similar flow estimates were computed at actual flow rates of 1–$8\ \mathrm{ml\ s^{-1}}$; linearity was excellent ($R > 0.999$, $m = 0.95$, $b = -0.03$), with an average underestimation of 5%.

3. Future work in speckle tracking

Research in several areas is required before ensemble tracking can become useful clinically. First, the grid

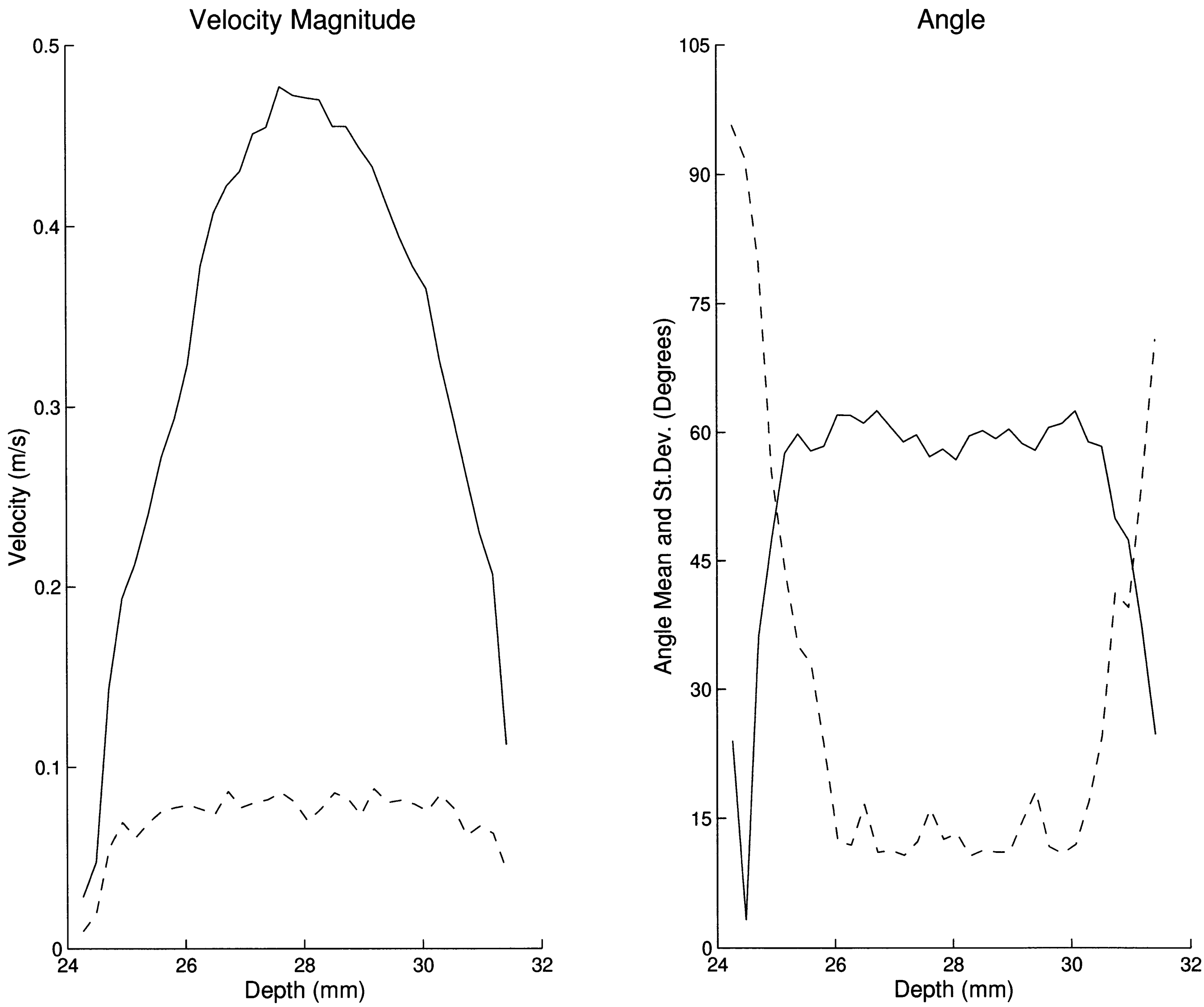

Fig. 2. Ensemble tracking measurement of flow in a laminar phantom with a nominal diameter of 7 mm, at a flow rate of 8 ml s^{-1}. The left panel shows the measured velocity magnitude profile (solid) and magnitude standard deviation (dashed). The right panel shows the measured angle profile (solid) and angle deviation (dashed). The kernel and search sizes were 1×20 and 2×40 pixels, respectively.

slopes interpolation method performs nearly ideally when the signal-to-noise ratio (SNR) is high. However, as the noise level increases, small translations are increasingly overestimated and large translations are increasingly underestimated. This problem is most prevalent in flow imaging, where the SNR is lower because of the relatively small amplitude of blood echoes relative to those from vessel walls and surrounding tissues. Contrast agents may provide one potential solution to this problem. Alternatively, we are investigating algorithm variations that will compensate for estimation bias that occurs when the SNR is low, without requiring contrast agents or a priori knowledge.

Similarly, a comprehensive evaluation of wall filters that will preserve blood echoes while sufficiently attenuating other signals is necessary. Such a study is critical for 2D speckle tracking because the lateral spatial frequencies from blood overlap with those from surrounding tissues, especially at low blood velocities.

Most of our work with ensemble tracking has utilized relatively high-frequency transducers (e.g. 7.5 MHz). However, lower frequencies will be required for cardiac applications. The shift to lower frequencies will result in wider lateral beamwidths, which will degrade lateral performance. Achieving accurate 2D velocity measurement will require that parallel beam symmetry, transmit energy, and the spacing between parallel beams be optimized for a given transducer and imaging range.

Translating the sub-pixel translations estimated by the grid slopes algorithm to physical distances requires knowledge of the spacing between parallel receive beams versus depth. Since this information cannot be gathered in vivo, an a priori estimation must be made. Because parallel receive processing results in a nonlinear spacing between beams versus depth [20], studies must determine how the beamspacing varies in various tissues for a given transducer and beamformer, and how these variations affect measurement accuracy.

One of our main clinical interests, quantifying flow in regurgitant cardiac jets, will require the use of lower frequency cardiac probes, as noted above, as well as the development of robust algorithms to compute flow from 2D velocity maps acquired from the proximal side of the jet [22]. The angle independence of 2D ensemble tracking may overcome the limitations that have prevented proximal flow quantification algorithms based on Doppler to gain widespread clinical use.

Finally, measurement of 3D velocities with ensemble tracking, which will be possible with the advent of 2D transducers and 3D imaging systems [3], may provide a major contribution to clinical flow and tissue motion assessment. How best to make such estimates, as well as how to display them to clinicians, will provide substantial challenges.

4. Summary and conclusions

Advancements in 2D speckle tracking made over the past decade have been outlined.

For bulk motion, tracking the movements of a larger speckle regions improves accuracy; however, when flow gradients are present, using larger kernels may degrade accuracy. In addition, early studies investigating the choice of RF or detected data agreed that RF data performed better; however, with limited axial sampling rates and high-frequency transducers, detected data can provide better performance. A simple SAD pattern matching algorithm was found to provide performance similar to normalized correlation, but at a substantially lower computational cost. For flow estimation, earlier methods based on line-by-line acquisition of kernel and search regions were limited because decorrelation caused by flow gradients made them unable to accurately quantify flow with axial components. Current ensemble tracking methods, based on parallel receive processing, minimize this decorrelation by tracking the motion of small kernels over smaller temporal and spatial intervals. Such methods have recently been shown capable of measuring flow with axial components. Finally, some of the challenges remaining for speckle tracking methods to gain clinical acceptance have been described.

Acknowledgement

We gratefully acknowledge Siemens Medical Systems, Inc. for loaning the Elegra scanner used in our laboratory. This work was supported by NIH grant R01-HL-50104.

References

[1] R.W. Gill, Measurement of blood flow by ultrasound: Accuracy and sources of error, Ultrasound Med. Biol. 11 (1985) 625–641.
[2] I.A. Hein, W.D. O'Brien, Current time-domain methods for assessing tissue motion by analysis from reflected ultrasound echoes – a review, IEEE Trans. Ultrason. Ferroelec., Freq. Contr. 40 (1993) 84–102.
[3] S.W. Smith, O.T. von Ramm, H.G. Pavy, High-speed ultrasound volumetric imaging system – part I: transducer design and beam steering, IEEE Trans. Ultrason. Ferroelec. Freq. Contr. 38 (1991) 100–108.
[4] L.N. Bohs, B.H. Friemel, B.A. McDermott, G.E. Trahey, A real time system for quantifying and displaying two-dimensional velocities using ultrasound, Ultrasound Med. Biol. 19 (1993) 751–761.
[5] I.A. Hein, 3-D flow velocity vector estimation with a triple-beam lens transducer – experimental results, IEEE Trans. Ultrason. Ferroelec., Freq. Contr. 44 (1997) 85–95.
[6] D.E. Robinson, F. Chen, L.S. Wilson, Measurement of velocity of propogation from ultrasonic pulse-echo data, 3-D flow velocity vector estimation with a triple-beam lens transducer – experimental results, Ultrasound Med. Biol. 8 (1982) 413–420.
[7] I. Akiyama, A. Hayama, M. Nakajima, Movement analysis of soft tissues by speckle patterns' fluctuation, JSUM Proc. (1986) 615–616. October.
[8] G.E. Trahey, J.W. Allison, O.T. von Ramm, Angle independent ultrasonic detection of blood flow, IEEE Trans. Biomed. Eng. 34 (1987) 965–967.
[9] B.S. Ramamurthy, G.E. Trahey, Potential and limitations of angle-independent flow detection algorithms using radio-frequency and detected echo signals, Ultrasonic Imaging 13 (1991) 252–268.
[10] E.J. Chen, I.A. Hein, J.B. Fowlkes, R.S. Adler, P.L. Carson, W.D. O'Brien, A comparison of the motion tracking of 2-D ultrasonic B-mode tissue images with a calibrated phantom, in: Proceedings IEEE Ultrasonics Symposium, IEEE, New York, 1991, pp. 1211–1214.
[11] D.I. Barnea, H.F. Silverman, A class of algorithms for fast digital image registration, IEEE Trans. Comput. C-21 (1972) 179–186.
[12] L.N. Bohs, G.E. Trahey, A novel method for angle independent ultrasonic imaging of blood flow and tissue motion, IEEE Trans. BME 38 (1991) 280–286.
[13] L.N. Bohs, B.H. Friemel, G.E. Trahey, Experimental velocity profiles and volumetric flow via two-dimensional speckle tracking, Ultrasound Med. Biol. 21 (1995) 885–898.
[14] B.H. Friemel, L.N. Bohs, K.R. Nightingale, G.E. Trahey, Speckle decorrelation due to two-dimensional flow gradients, IEEE Trans. Ultrason. Ferroelec. Freq. Contr. 45 (1998) 317–327.
[15] K. Ferrara, V.R. Algazi, A theoretical and experimental analysis of the received signal from disturbed blood flow, IEEE Trans. Ultrason. Ferroelec. Freq. Contr. 41 (1994) 172–184.
[16] D.P. Shattuck, M.D. Weinshenker, S.W. Smith, O.T. von Ramm, Explososcan: A parallel processing technique for high speed ultrasound imaging with linear phased arrays, J. Acoust. Soc. Am. 75 (1984) 1273–1282.
[17] L.N. Bohs, B.J. Geiman, M.E. Anderson, S.M. Breit, G.E. Trahey, Ensemble tracking for 2D vector velocity measurement: experimental and initial clinical results, IEEE Trans. Ultrason. Ferroelec. Freq. Contr. 45 (1998) 912–924.
[18] O. Bonnefous, Measurement of the complete (3D) velocity vector

of blood flows, In: Proceedings IEEE Ultrasonics Symposium, IEEE, New York, 1988, pp. 795–799.
[19] B.J. Geiman, L.N. Bohs, M.E. Anderson, S.M. Breit, G.E. Trahey, A novel interpolation strategy for estimating sub-sample speckle motion, Physics Med. Biol. (2000) in press.
[20] L.N. Bohs, S.C. Gebhart, M.E. Anderson, B.J. Geiman, G.E. Trahey, 2D motion estimation using two parallel receive beams, IEEE Trans. Ultrason. Ferroelec. Freq. Contr. (2000) submitted.
[21] M.E. Anderson, L.N. Bohs, S.C. Gebhart, A comparison of flow tracking techniques: spatial quadrature with phase-sensitive axial demodulation versus speckle tracking, IEEE Trans. Ultrason. Ferroelec. Freq. Contr. (2000) in press.
[22] P.M. Vandervoort, D.H. Thoreau, J.M. Rivera, R.A. Levine, A.E. Weyman, J.D. Thomas, Automated flow rate calculations based on digital analysis of flow convergence proximal to regurgitant orifices, J. Am. Coll. Card. 22 (1993) 535–541.