Chapter 26

Introduction to speckle tracking in cardiac ultrasound imaging

Damien Garcia¹, Pierre Lantelme¹,² and Éric Saloux³

In this chapter, we will first recall some basic principles of speckle tracking. The fundamentals of speckle tracking in a wider context are essentially described in Chapter 13. We will then treat speckle-tracking echocardiography and echocardiographic particle image velocimetry (echo-PIV) and indicate a number of clinical applications in the context of evaluation of cardiac function. We will then briefly introduce color Doppler approaches complementary to speckle tracking. We will finally present how speckle-tracking techniques could benefit from high-frame-rate echocardiography (also called “ultrafast echocardiography”). We will conclude with the expected contribution of high-frame-rate ultrasound for speckle tracking in three-dimensional (3-D) echocardiography.

26.1 Speckle formation and speckle tracking

The word “speckle” refers to the granular appearance of an image generated by a coherent imaging system, such as laser, optical coherence tomography, and ultrasound. As explained in detail in Chapter 2, speckles appear when a random collection of scatterers is illuminated by waves whose wavelength is larger than the size of the individual scatterers. The grainy aspect of a speckle pattern is produced by the multiple backscattered signals of similar frequency that interfere constructively and destructively, depending on their relative phases and amplitudes (Figure 26.1). In medical ultrasound imaging, as soft tissues contain many scatterers, the ultrasound waveforms detected by the transducer are the combination (interference) of the different wave reflections induced by the distinct scatterers. The resulting speckles are visible in the unfiltered gray-level (B-mode) images as dark and bright specks.

¹INSERM. CREATIS, UMR 5220, U1206, Université de Lyon, INSA-Lyon, Université Claude Bernard Lyon 1, UJM-Saint Étienne, France
²Fédération de Cardiologie Croix-Rousse, Lyon-Sud, Hospices Civils de Lyon, Université de Lyon, France
³Department of Cardiology, CHU de Caen. EA 4650, SEILIRM, Université de Normandie, France
When the individual contributions of the scatterers are independent, the speckle patterns can be accurately modeled by various statistical distributions whose physical meanings have been thoroughly discussed in [1]. Chapter 3 also provides an overview of the statistical models introduced since the pioneering work of Burkhardt in 1978 [2]. Because medical ultrasound images are made exclusively of speckles, they can significantly affect visualization quality or postprocessing tasks, and in turn negatively impact the diagnostic potential of medical ultrasound [3]. Despeckling can thus be a necessary task in some specific imaging situations. This angle is a main theme of this handbook and is largely addressed in the chapters devoted to speckle filtering (i.e., part II). On the contrary, in the present chapter, we consider the speckle patterns as intrinsic signatures of the insonated medium. In such a case, these distinctive imprints must be sufficiently preserved from one frame to the next one to allow analysis of tissue dynamics through speckle tracking.

Speckle tracking for ultrasound imaging has been introduced by Trahey et al. [4], from Duke University, North Carolina, United States, to produce a blood flow velocity vector field in a human vein (Figure 26.2, left). Although it does not appear in the paper’s references, it is likely that Trahey et al.’s approach has been influenced by two-dimensional (2-D) speckle velocimetry, a former speckle photography technique to measure 2-D velocity fields in unsteady flows [6]. Since then, speckle tracking in medical ultrasound imaging has been the subject of a yearly increasing number of investigations (Figure 26.3), mainly in the field of myocardial strain imaging (deformation imaging of the cardiac muscle). Another strong interest for speckle tracking later emerged in contrast echocardiography (cardiac ultrasound imaging with contrast agents) to display 2-D blood velocity vector fields in the cardiac left ventricular cavity, a technique often called “echocardiographic particle image velocimetry” (or echo-PIV). The following paragraphs will describe these two common applications of speckle tracking in cardiac ultrasound imaging. Although strain imaging and echo-PIV are also of interest in ultrasound vascular imaging, the vascular field will not be discussed in the present chapter. We invite the
Figure 26.2  First applications of speckle tracking. Left: Venous blood velocity profile obtained with a cross-correlation-based block-matching approach [4]. Right: Myocardial motion from a global Horn–Schunck optical flow method [5]

Figure 26.3  Yearly occurrence of “speckle tracking” in abstracts and/or titles of MEDLINE-referred papers. The first occurrence of “speckle tracking” is in [7]
interested reader to refer to the chapters that address these specific topics. One of the reasons for focusing on cardiac imaging is that speckle tracking is mostly used clinically in the context of cardiac evaluation.

**In summary**—Speckles are issued from the interference of the wave reflections induced by the tissue scatterers. Although considered as noise in some imaging applications, speckle patterns also represent local signatures of the insonified tissues. These speckles can be tracked to determine frame-to-frame motion.

### 26.2 Basic principles of speckle tracking

As explained earlier, speckles can be considered as acoustic markers of the insonified tissues. In a time series of ultrasound images, these markers are sufficiently preserved from one frame to the next if the frame rate is high enough. In this latter condition, it is therefore possible to locally track the speckle patterns and thus deduce the local tissue displacements with a frame-wise approach as explained in detail in Part III. For example, in the current clinical practice, a frame rate of 50–80 frames/s is recommended to obtain optimal conditions for speckle tracking in the resting heart [8]. Ultrasound speckles can be tracked frame to frame by various approaches, such as differential optical-flow methods [9,10] or block-matching algorithms [10–12]. A block-matching algorithm is a method used to estimate motion in a video sequence by locating similar blocks between two successive images. It is generally well adapted to retrieve relatively large frame-to-frame displacements. A simple though standard block-matching method is to compare the intensities of the pixels using the sum of absolute/squared differences measure [13]. Among a number of similarity criteria, the normalized cross-correlation was historically the first [7], and is still one of the most applied methods, in medical ultrasound imaging [14–17]. In the following, although different similarity criteria can be used, we thus focus on the normalized cross-correlation without loss of generality. The normalized cross-correlation can be evaluated directly in the spatial domain by using small subwindows in one frame, and search areas of larger size in a subsequent frame [18]. Another possibility is to calculate the normalized cross-correlation in the Fourier domain (Figure 26.4), a consequence of the Wiener–Khinchin theorem [19]. The latter approach is often called “phase correlation” [20]. The peak location of the normalized cross-correlation corresponds to the local displacement with a pixel precision. A subpixel precision can be returned from 2-D fitting of the correlation peak [19]. In its simplest form, speckle tracking by cross-correlation can be summed up by the following three-step process: (1) division of two successive B-mode images into small subwindows, (2) normalized cross-correlation of the subwindow pairs, and (3) peak fitting and estimation of the displacements. In the Fourier domain (phase correlation), it boils down to the following steps (see also Figure 26.4):

Let $I_1$ and $I_2$ represent two successive gray-level B-mode images. These two images are both subdivided into evenly-spaced subwindows of size $(m \times n)$, $w^1_k$ and
Figure 26.4 Speckle tracking using the normalized cross-correlation implemented in the Fourier domain. This block-matching scheme can be generalized with other similarity criteria, such as the SAD or SSD (sum of absolute/squared differences)

\[ w_k^1, \text{ with } k = 1 \ldots M, M \text{ denoting the total number of subwindows. Size and overlap of the subwindows must be adapted to adjust the precision/accuracy compromise, as well as the resolution of the output displacement field. Let } W_k^1 = \mathcal{F}(w_k^1) \text{ and } W_k^2 = \mathcal{F}(w_k^2) \text{ be the 2-D output displacement field. The Fast Fourier Transform (FFT)-based normalized cross-correlation for each subwindow } k \text{ is given by} \]

\[ \text{NCC}^k = \mathcal{F}^{-1}\left(\frac{w_k^1 w_k^2}{|w_k^1 w_k^2|}\right). \]

The inverse Fourier transform is denoted by \( \mathcal{F}^{-1} \), and the overbar denotes the complex conjugate. The divisions and multiplications are elementwise. The relative translation \((\Delta x^k, \Delta y^k)\) between the two subwindows \(w_k^1\) and \(w_k^2\) is given by the location of the peak in \(\text{NCC}^k\):

\[ (\Delta x^k, \Delta y^k) = \arg \max_{(i,j)} \left(\text{NCC}^k\right) \]

To determine the translation with subpixel accuracy, a simple and robust method is to fit the correlation peak to some function, such as a paraboloid or Gaussian surface [19]. The displacement vectors in standard units are finally deduced by knowing the pixel size.

The phase correlation is only one approach for speckle tracking among a large variety of template-matching methods; it was here described in its most basic form. Part III provides a broader overview of the principles of speckle tracking. More advanced numerical methods exist in the technical literature devoted to image registration or optical PIV. Just to name a few for cross-correlation, these methods include ensemble cross-correlation, coarse-to-fine analysis, and interrogation subwindow deformations [19]. Other algorithms can also be found, for example, in the families of differential optical flow [9,21,22], nonrigid transformation [23,24], and point matching [25–27]. Yet, speckle tracking in medical ultrasound imaging is traditionally associated with cross-correlation, likely for historical and expedient reasons. On a final note, regardless of the approach used to track the speckles, robust postprocessing of the raw displacements is essential if differential quantities.
must be derived (e.g., strains, shears), as required in strain imaging. Such post-processing can include replacement of incorrect data and smoothing of the vector field. To this effect, advanced algorithms exist in the literature [28,29,16]. Post-processing has a major impact on the output vector field and on the derived differential quantities. In the commercially available workstations for ultrasound speckle tracking, postprocessing is hidden in “black boxes” and differs from vendor to vendor. As discussed later, intervendor variability is the main reason why global longitudinal strain (GLS) is the only myocardial strain parameter which may be used in routine clinical practice [30].

In summary—Numerous algorithms have been derived for speckle tracking in ultrasound imaging. Both local and global techniques have been introduced. One of the most used approaches is local block matching based on the measure of the normalized cross-correlation.

26.3 Speckle-tracking echocardiography

The vast majority of the “speckle tracking” references included in the MEDLINE database (Medical Literature Analysis and Retrieval System Online, National Library of Medicine—National Institutes of Health) are devoted to the motion analysis of the left ventricular myocardium (Figure 26.5). This trend has emerged naturally since echocardiography is the main clinical imaging modality for the evaluation of cardiac function. As the temporal resolution of echocardiography became satisfactory through the 1980s, there was an increasing interest in the quantitative analysis of the myocardium movement, since its visual evaluation was subjective and thus highly operator-dependent. In particular, the first thorough

Figure 26.5 “Tissue Doppler” vs. “speckle tracking” for strain imaging. Yearly occurrence of (“tissue Doppler” or “speckle tracking”) and “strain”) in abstracts and/or titles of MEDLINE-referred papers
investigations worth mentioning are those of Mailloux et al., from École Polytechnique de Montréal, Canada [5,9]. In their studies, the authors estimated the cardiac motion field in a short-axis view (i.e., a cross-sectional slice of the heart, see right picture in Figure 26.2) by tracking the speckle patterns automatically using the Horn–Schunck’s global optical flow method [31]. Later on, block-matching schemes have been preferred and are now parts of commercial software [32,33]. Speckle tracking, in and of itself, returns local displacements and/or velocities and thus truly characterizes wall motion. To distinguish active myocardial motion from passive translational or tethering movements, physicians have preferably examined the regional deformations, mostly by estimating regional strains and strain rates. This echocardiographic modality for regional deformation imaging is referred to as “strain imaging.” Strain (deformation) and strain rates (rate of deformation) can be derived from the spatial derivatives of the displacements and velocities, respectively. Strain (rate) imaging originated from tissue Doppler [34–36], an echocardiographic technique that uses the Doppler mode to measure the velocities of the cardiac muscle (myocardium). Although tissue Doppler was the first modality of choice for cardiac strain imaging, speckle tracking in gray-scale 2-D images has then become the standard procedure since 2010 (Figure 26.5). In particular, a strong interest in cardiac speckle tracking has been demonstrated in cardiac resynchronization therapy [37]. When speckle tracking is used in cardiac imaging, strain imaging is commonly referred to as “speckle-tracking echocardiography.” The advantages of speckle tracking over tissue Doppler for myocardial strain imaging have been the subject of key handbooks and review papers [38–40]. To enumerate a few: speckle tracking is angle independent, offers better spatial and temporal resolutions, and is less sensitive to noise. In this chapter, we do not dwell upon this particular point.

Over the last 10 years, a number of clinical software tools and algorithms have been marketed by different vendors for speckle-tracking echocardiography. The access of speckle-tracking echocardiography presented a new opportunity for a better evaluation of the heart function. Numerous clinical studies show that speckle-tracking echocardiography can provide complementary quantification of regional and global cardiac function. More specifically, speckle-tracking echocardiography offers a unique insight into the impairment of the left ventricular function. Some of its main areas of clinical application are [41]: detection of subclinical myocardial dysfunction, diagnosis of ischemia and location of myocardial infarction, risk stratification in cardiomyopathy, prediction of the response to cardiac resynchronization therapy, assessment of the systolic and diastolic dysfunctions, evaluation of myocardial mechanics in heart failure, follow-up after heart valve surgery. Readers interested in a detailed portrayal of the clinical applications of speckle-tracking echocardiography are referred to recent clinical review papers [41–43]. Despite its broad clinical appeal over the last decade (Figure 26.5) and the numerous studies described in high-impact journals, no clear consensus has yet been reached to standardize left ventricular strain imaging [8]. The main reason for this reluctance has been the significant variability in regional strains, which is observed among the software packages. Intervendor inconsistencies stem from
different causes: (1) speckle tracking is dependent upon the image characteristics, which differ across ultrasound scanners; (2) divergences in the terminology describing the myocardial mechanics are also noted; and (3) finally, and it is likely the primary source of discordances, the software packages use proprietary algorithms for pre- and postprocessing, speckle tracking, and data regularization. To complicate matters, whether the whole process is more or less unsupervised can also affect interoperator reproducibility [8,30]. As a consequence, the current echocardiographic guidelines do not recommend quantitative measures of regional deformation, despite the strong clinical interest of strain imaging. Conceding the critical necessity for consistencies in speckle-tracking echocardiography, leaders of the European Association of Cardiovascular Imaging and the American Society of Echocardiography have invited technical representatives from several industrial partners to cooperate with a view to reducing intervendor variability in strain imaging [8,44]. The main conclusion from this task force is that global longitudinal strain (GLS) is the most robust deformation parameter and is presently the only myocardial strain parameter which may be safely used in routine clinical practice [30]. GLS is a diagnostic and prognostic marker of the global left ventricular systolic function, which can be measured clinically by speckle-tracking echocardiography [45]. It reflects the relative longitudinal contraction (in %) of the left ventricular myocardium (Figure 26.6). The instantaneous GLS (in %) can be written as \( \text{GLS}(t) = 100 \times \frac{L(t) - L(ED)}{L(ED)} \), where \( L(t) \) is the longitudinal myocardial length at time \( t \), and \( L(ED) \) is the length at end-diastole. The GLS peak is around −20% in normal subjects [46]. Speckle-tracking echocardiography will receive a renewed resurgence of interest if manufacturers can match their regional strains. A plan of action will consist in comparing echo-derived strains with local strains determined by sonomicrometry in an in vitro phantom [47] which reproduces myocardial shortening, torsion, and contraction (lengthening, untwisting, and dilation) during systole (diastole).

**In summary**—Speckle-tracking echocardiography has supplanted tissue Doppler for strain (rate) imaging. Although speckle-tracking echocardiography was first designed to determine regional deformations, it was found that substantial intervendor variability prevents its use in routine clinical practice. It followed that the GLS is presently the only myocardial strain parameter which may be safely used. Speckle-tracking echocardiography, which is a local tool, is thus constrained to limit itself to global assessment.

**26.4 Clinical utility of global longitudinal strain in speckle-tracking echocardiography**

Assessment of global left ventricular systolic function plays a key role in the prognosis of cardiac diseases. The most widely used parameter is the left ventricular ejection fraction, commonly determined by echocardiography.
Left ventricular ejection fraction represents the amount of blood leaving the left ventricle each time it contracts. An ejection fraction of 60% means that 60% of the total amount of blood within the left ventricle is pumped out with each heartbeat. A normal ejection fraction is between 50% and 70%. It can be calculated by delineating the endocardium in 2-D echo images, at both end diastole and end systole [48]. Although widely used, left ventricular ejection fraction has a number of important limitations in assessing systolic function and can offer a poor prognosis in many situations. In comparison, the prognostic value of the abovementioned GLS is recognized to be superior [45] since GLS is more sensitive to detect subtle changes in myocardial function (Figure 26.6). In addition, some cardiomyopathies can be marked by a reduced GLS although the ejection fraction is preserved. We here briefly describe the clinical utility of GLS in a few clinical contexts.

**Cardiotoxicity in chemotherapy**—Breast cancer is the principal type of cancer in women. Chemotherapeutic treatments (administration of anticancer drugs) increase rate of cure and reduce relapse significantly. Chemotherapy, however, is limited by the risk of cardiotoxicity, which can appear early in the treatment and may induce irreversible heart failure if not detected subclinically. Treatment must be promptly discontinued or modified before the onset of observable symptoms related to cardiotoxicity. Cardiac dysfunction induced by chemotherapy is commonly diagnosed through echocardiographic examination, primarily on the basis of the left
ventricular ejection fraction. In addition to ejection fraction, it is now recommended to measure GLS [49], as an early reduction in GLS (Figure 26.6) may help to diagnose subclinical systolic dysfunction before a drop in ejection fraction [50,51].

*Aortic stenosis*—Aortic stenosis refers to narrowing of the aortic valve opening, which restricts blood ejection [52]. The most common cause is senile calcification of the valve cusps, making the valve stiff and obstructive. If the stenosis is severe, the aortic valve must be replaced surgically or transcutaneously. Timing for replacement is decided through the presence of symptoms, the severity of the stenosis, and its impact on the left ventricular volume. A substantial reduction in ejection fraction may denote irreversible damage of the myocardium. Left ventricular GLS has been shown to provide incremental prognosis in aortic stenosis and to predict mortality in patients with preserved ejection fraction [53,54]. Incorporation of GLS into the echocardiographic parameters for the assessment of aortic stenosis may thus improve the determination of the optimal timing for aortic valve replacement [55].

*Coronary artery disease and myocardial infarction*—Peak GLS is significantly impaired in patients with myocardial infarction and correlates with infarct size [56,57]. A significantly reduced GLS (<15%; the normal value being 20%) has also been shown to be an independent predictor of sudden cardiac death after acute myocardial infarction, even if left ventricular ejection fraction is preserved [58,59]. GLS impairment can also be effectively used to detect non-obstructive coronary heart disease [60], and subclinical left ventricular systolic dysfunction in patients with obstructive coronary heart disease [61].

Left ventricular GLS derived by speckle tracking has also been shown to be clinically relevant in terms of prognosis in patients with hypertrophic cardiomyopathy [62], mitral valve regurgitation [63], and aortic regurgitation [64]. On one final note, the GLS is also of clinical value in evaluation of right ventricular function [65].

In summary—Left ventricular ejection fraction has long been the most essential parameter for global assessment of left ventricular function. However, GLS might have a better prognostic value in a variety of cardiac conditions. Left ventricular ejection fraction and GLS are thus complementary clinical indices of cardiac function. Because GLS has been shown to be reproducible despite intervendor variability in strain imaging, it is likely that GLS will be soon part of the routinely measured parameters during an echocardiographic examination.

26.5 Echocardiographic particle image velocimetry (Echo-PIV)

When speckle tracking is applied to contrast-enhanced blood images, the technique is generally called “echo-PIV,” the short name for “echocardiographic particle image velocimetry” [66]. Echo-PIV is an adaptation of optical PIV, a commonly
used indirect technique for measuring two or three velocity components in experimental fluid dynamics [67]. In some way, echo-PIV is a resurgence of speckle velocimetry [6], from which optical PIV emerged. In optical PIV, two sequential laser pulses illuminate a slice or volume of a flow seeded with microscopic light-scattering particles. The scattered light is recorded in two successive images by digital camera(s). These images are subdivided into small areas for calculating the mean particle displacement between two corresponding subareas using block-matching techniques (as in Figure 26.4). Knowing the lag between two laser pulses, the particle velocities can be thus determined [68]. Since visible laser lights are used, optical PIV requires transparent experimental setups. With the objective of obtaining velocity vector fields in the blood circulation, optical PIV was adapted to ultrasound imaging [7,69,70] to become echo-PIV. In echo-PIV, well-established speckle-tracking methods (such as block matching through normalized cross-correlation, see Section 26.2) are applied on contrast-enhanced B-mode images obtained during intravenous administration of gas-filled microbubbles (contrast agents). Ultrasound contrast agents are made of gas encapsulated in a hydrophilic shell. Their mean size (<6 μm) is less than the diameter of the pulmonary and systemic capillaries they must transit when delivered intravenously [71]. In contrast to optical PIV where Mie scattering dominates, Rayleigh scattering governs echo-PIV. In other words, while optical PIV generates images of the particles, echo-PIV does not produce images of the microbubbles but images of the interferences (speckles). Echo-PIV has been successfully used to investigate the carotid blood flow [72,73]. As with strain imaging, however, echo-PIV has spread clinically in transthoracic echocardiography, the most common application being the quantitative analysis of the main intraventricular vortex (Figure 26.7) for a better evaluation of left ventricular cardiac function [74–76], as clarified in the next section.

Figure 26.7  Echo-PIV in the left ventricle. Echo-PIV makes it possible to decipher and quantify the dynamics of the vortex that forms in the left ventricle during left ventricular filling. Tracking of the intraventricular microbubbles were here carried out in a three-chamber view. Dynamical properties of this vortex are potential biomarkers of the diastolic function. Courtesy of Pr. Y.J. Moon, School of Mechanical Engineering, Korea University
In summary—Echo-PIV is a promising tool to decipher the blood flow dynamics within the left ventricle. It requires administration of gas-filled microbubbles to enhance the blood signal for blood speckle tracking, which may prevent its routine clinical applicability.

26.6 Potential clinical utility of vortex flow imaging in echo-PIV

Left ventricular diastole refers to blood filling of the left ventricle. An accurate assessment of left ventricular diastolic function is of utmost clinical importance in patients with dyspnea (shortness of breath) or heart failure. Heart failure occurs when the heart muscle is weakened and is unable to pump enough blood to meet the body’s needs for metabolites and oxygen. Among these patients, about half of them have normal or nearly normal left ventricular ejection fraction. Diagnosis of diastolic dysfunction can be ambiguous since patient’s history, physical examination, electrocardiogram, or chest X-ray is frequently unhelpful. The wide accessibility of echocardiography, as well as its ability to provide real-time information, makes ultrasound the prerequisite technique for evaluating left ventricular diastolic function. However, current echocardiographic indices of diastolic function have major limitations that may hinder accurate diagnosis. Recent studies suggest that a deeper look at the intraventricular flow during filling can improve diastology (assessment of diastolic function) by echocardiography [77,78]. The intraventricular flow dynamics is indeed much richer during diastole (filling phase) than during ejection. Diastole of the left ventricle is featured by the formation of a wide whirling motion (vortex) which originates during early filling [79]. In the normal heart, most of the left ventricular intracavitary blood volume is involved in the vortical pattern [80]. Clinical observations in patients and normal subjects indicate that the diastolic intraventricular blood vortices have particular shapes and locations, which can be biomarkers of the cardiac function [74,80,81–83]. Characterization of the intraventricular vortex flow by echocardiography is becoming increasingly popular in the clinical literature [77,78], and it is probable that measures of the intracardiac vortex hemodynamics can play a key role in the assessment of cardiac function (Figure 26.7). Vortex formation in the normal left ventricle was first outlined by Rodevand et al. who interpreted intraventricular color Doppler velocities and pulsed-wave Doppler spectra [84]. The two predominant echocardiographic tools for vector flow imaging within the left ventricle are now echo-PIV [85,76] and “vector flow mapping” [86,87], a technique based on color Doppler imaging (see Section 26.7). Figure 26.7 shows how echo-PIV can disclose the vortex formation in the left ventricle.

Since vortex flow imaging by ultrasound imaging is a very recent technique, no clear clinical conclusion has yet been reached on the vortical parameters that must be preferably determined (size, vorticity, circulation, energy, elongation, time, location . . . ) to improve the assessment of cardiac function. In vivo research
on this topic is exploratory, and the pathophysiological relevance of intraventricular fluid dynamics still needs to be demonstrated in large cohorts. In addition, only a very few teams have investigated the potential clinical relevance of vortex flow imaging by echo-PIV, so that intergroup reproducibility has not yet been explored. However, it appears that diastole has to be more than a mere passive suction: the dynamics of the diastolic vortex has a key physiological impact on left ventricular filling. Blood flow keeps swirling in late diastole and maintains momentum during the isovolumetric contraction phase [88]. This avoids any period of blood flow stasis and helps coupling filling phase to ejection phase. Broadly speaking, the diastolic vortex can be seen as a primer of the cardiac pump. This diastole-to-systole dynamical coupling has been shown to be suboptimal in patients with heart failure and reduced left ventricular ejection fraction [89]. In line with these observations, echo-PIV investigation in patients with acute myocardial infarction revealed a progressive decline in vortex flow dynamics as left ventricular dysfunction develops [90]. Vortex geometry and dynamics, as well as intraventricular blood transport, have also been shown to be modified after left ventricular remodeling, i.e., after changes in cardiac size and shape [74,76,91]. We invite the reader to consult recent review papers for a clinical survey of intraventricular flow imaging by echocardiography [76–78,92]. The principal clinical limitation of echo-PIV is the requirement of intravenous administration of microbubbles. Although no major side effect has been noticed, this procedure is time and staff consuming and thus cannot be recommended in routine clinical practice. To put an end to this obstacle, Fadnes et al. introduced a contrast-free procedure to track the native speckles of blood [93]. This speckle-tracking tool was shown to offer better visualization and quantification of septal defect—a congenital heart disease in which there is an opening in the septum—in neonates [94]. It has been proposed in the context of high-frame-rate echocardiography (see Section 26.8) and adapted for transthoracic echocardiographic in the adults (Figure 26.8).

**In summary**—The physiological and pathophysiological implications of intraventricular fluid dynamics are still being prospected. Its analysis by echo-PIV can bring additional insights into the echocardiographic evaluation of diastolic function. As was (and still is) the case for speckle-tracking echocardiography, large-scale clinical trials are required to demonstrate the significance of echo-PIV in terms of cost effectiveness and clinical outcomes before being accepted as a standard echocardiographic examination.

### 26.7 Color Doppler as an alternative or complementary to speckle tracking

As shown in Figure 26.5, tissue Doppler had been the most frequent method for strain and strain rate imaging before 2010. One key advantage of tissue Doppler is
that it directly yields Eulerian velocities of myocardial or blood tissues. One main drawback is its angle dependency since Doppler modes return the axial velocity components, i.e., the components parallel to the ultrasound beam axes [40]. Speckle tracking thus logically imposed itself as the superior method for motion analysis of the left ventricle. Likewise, as discussed in the previous section, speckle tracking has become the most accepted technique for investigation of the intraventricular blood flow. To work around the one-dimensional issue of color Doppler, multi-Doppler approaches have been developed. By registering two series of color Doppler images acquired with significantly different steering angles, Arigovindan et al. produced long-axis motion fields of the left ventricular myocardium [95]. Gomez et al. then generalized this method [96,97] and proposed a time-resolved volumetric reconstruction of the intraventricular flow from volumetric color Doppler images (Figure 26.9, left picture). Postprocessing of conventional color Doppler images has also been proposed to retrieve 2-D intraventricular vector fields in apical long-axis views [86,98]. The vector flow mapping technique introduced by Garcia et al. (Figure 26.9, right picture) is now implemented in a commercially available ultrasound machine [99] and has been the basic tool for a number of clinical pilot studies [87,100]. Another very promising strategy consists in combining the better of the two worlds in an optimization problem [101–103]. Since this approach requires both color/tissue Doppler and B-mode, it will be best suited for high-frame-rate ultrasound imaging. With the advent of high-frame-rate echocardiography

Figure 26.8  Contrast-agent-free echo-PIV overlaid on color Doppler. Intraventricular vortex revealed by speckle tracking without administration of contrast agent. It was overlaid on color Doppler to make the flow visualization easier (red ¼ towards the probe, blue ¼ away from the probe). Courtesy of Pr. Lasse Løvstakken, Norwegian University of Science and Technology
Improvements in motion detection can be expected by getting both color/tissue Doppler and B-mode images in a single heart beat at high frame rates [104]. By using advanced and robust algorithms, the inter-software variability may ultimately become small enough to allow accurate estimation of regional deformations.

**In summary**—Color Doppler can be combined to speckle tracking to improve motion tracking. Although this assertion remains speculative, some evidence of such potential improvement has been reported in the recent literature.

### 26.8 Potential benefits of high-frame-rate echocardiography

Conventional echocardiography consists in scanning the heart using a series of successive focused beams (typically 64 to 128) that cover the sector of interest (see Figure 26.10, leftmost picture). The resulting scanlines are then stacked together to reconstruct a single image. The time required to build one frame is thus proportional to the number of gathered lines. Since the acoustic waves must travel down to the maximal depth then come back to the probe at the speed of $\sim 1,540 \, \text{m/s}$, wave travel time is roughly 0.2 ms in adult echocardiography (when considering a maximal range of 15 cm). For a wide sector containing 100 scanlines, the frame...
rate is thus around 50. Parallel computing, high-performance data transfer and high-speed processors have recently changed ultrasound imaging drastically [105–107]. Instead of transmitting a series of successive focused emissions (Figure 26.10, leftmost picture), large planar or circular wavefronts (Figure 26.10, third and fourth pictures) can be emitted to insonify wide regions. Alternatively, several focused beams can be transmitted simultaneously (Figure 26.10, second picture). The radiofrequency echoes are acquired all together by each element to reconstruct an image in postprocessing, which offers high-frame-rate (or ultrafast) ultrasound imaging. In comparison with conventional echocardiography, it has been shown in normal subjects that high-frame-rate echocardiography can provide a 5-fold increase in frame rate without deteriorating image quality significantly [108,104].

In the current clinical practice, a useful rule of thumb is that the frame rate must be roughly equal to the heart rate to yield optimal conditions for speckle-tracking echocardiography. In some echocardiographic examinations, such as during the evaluation and management of coronary artery disease, it can be clinically relevant to increase the heart rate, a technique called “stress echocardiography.” It consists in scanning a patient during a cardiac stress induced by exercise (treadmill or bicycle) or by administration of a pharmacological stressor. The images acquired before and after stress are compared to detect the presence of regions with abnormal wall motion [109]. Dobutamine is a widely available stressor which increases heart rate up to 120–140 beats per minute. In a stressed myocardium, the mechanical events become shorter; the acquisition frame rate thus should be increased proportionally with the heart rate (>100 frames per second) according to the abovementioned rule of thumb. Acquiring the whole left ventricular myocardium at such high frame rates can be very challenging with the conventional imaging systems [110]. The physicians have several stratagems to increase the frame rate, decrease imaging depth, narrow sector size, and reduce line density. However, this is obviously at the expense of image quality, which can

Figure 26.10 Conventional vs. High-frame-rate echocardiography. Conventional echocardiography requires ~100 scanlines to produce a wide-sector cardiac image. High-frame-rate echocardiography can provide a 5-fold increase in frame rate without deteriorating image quality significantly. See also [105]
make speckle tracking difficult or impossible. Speckle-tracking echocardiography under dobutamine-induced stress could thus benefit from high-frame-rate ultrasound imaging, where frame rates greater than 250 with well-preserved speckles have been reported in the left ventricle [104]. Up to now, most of the speckle-tracking echocardiography studies have been devoted to single cardiac chamber (mostly the left ventricle), for several practical reasons. A recent study performed in normal subjects, however, illustrates that simultaneous measurement of longitudinal strains in all four cardiac chambers could provide new insights into inter-chamber functional relationships [111]. Since simultaneous 4-chamber strain requires both large depth and width, it is very likely that high-frame-rate echocardiography can also enlarge the spectrum of its applications. Note finally that the frame-rate limitation is also especially true with echo-PIV, even in the resting heart, since greater displacements (intraventricular blood flow) must be measured. In conventional echo-PIV, the frame rate can be increased through decreasing depth and narrowing sector. Echo-PIV could thus take advantage of high-frame-rate echocardiography. An example to this effect is given in Figure 26.8.

**In summary**—High-frame-rate echocardiography is an emerging modality in clinical imaging. It can provide very high frame rates without degrading image quality significantly. Dobutamine stress echocardiography or full-heart strain imaging may profit from this innovative imaging method.

### 26.9 Toward volumetric speckle-tracking echocardiography

3-D echocardiography is a recent and major innovation in cardiac ultrasound. It is a volumetric method of visualizing the dynamic anatomy and function of the heart. Over the last years, there has been a steady stream of fundamental and clinical publications on 3-D echocardiography [112]. One can safely advance that 3-D transthoracic echocardiography would not have had the rise in popularity that we observe today if it had not benefited from 3-D transesophageal echocardiography. 3-D transesophageal echocardiography has made it possible to see the previously invisible, in particular, the mitral valve. It marked the beginning of modern catheter-based interventions [113] such as mitral valve repair, left atrial appendage occlusion, closure of paravalvular leaks and septal defects, and transcutaneous aortic valve implantation. In this interventional context, 3-D transesophageal echocardiography allows visualizing the entire percutaneous procedures in a single volumetric view. 3-D transthoracic echocardiography was first left behind for two main reasons: (1) 2-D transthoracic echocardiography is fast and good enough and (2) the off-line analysis was laborious due to limited software packages. Owing to the recent progress of user interfaces and the related postprocessing analysis, the advent of 3-D transthoracic echocardiography has significantly impacted the clinical management of cardiac diseases. Clinicians now agree that 3-D echo imaging
can be both complementary and supplementary to 2-D echocardiography. A comprehensive description of practical applications of real-time 3-D echocardiography can be found in the recent book of Buck et al. [114]. For example, 3-D transthoracic echocardiography can allow the physician to obtain accurate estimates of the cardiac chamber volumes (Figure 26.11). This imaging modality, however, is still rarely used at patient bedside. With up-to-date clinical scanners, it is indeed not possible to perform volumetric sequences at adequate temporal and spatial resolutions in a single heartbeat. To obtain volume images at acceptable spatial resolution, they must be captured by stacking several narrow volumes acquired during successive heart cycles, typically 5 for a left ventricle. This technique requires electrocardiogram (ECG) gating, patient’s breath-holding, and negligible beat-to-beat variability. On the other hand, single-beat 3-D imaging can be achieved in specific situations, but at the expense of spatial resolution. Clinical benefits of 3-D transthoracic echocardiography have been demonstrated in the evaluation of (1) cardiac

Figure 26.11 Three-dimensional echocardiography. 3-D echocardiography makes it possible to get an accurate estimate of the cardiac chamber volumes. Volumetric speckle-tracking echocardiography, however, is far from being readily available.
chamber volumes and masses, (2) regional left ventricular wall motion, and (3) cardiac shunts and heart valve regurgitations. Due to its technical limitations, however, 3-D echocardiography is very challenging, if not infeasible, in patients with cardiac arrhythmias and/or breath-holding difficulty. To become a standard echo examination procedure, 3-D echocardiography must offer high spatial and temporal resolution, preferably in a single heartbeat. It is thus obvious that the path toward reproducible volumetric speckle-tracking echocardiography is still strewn with major, but surmountable, obstacles. Efforts were already made with a synthetic database of volumetric and realistic B-mode image sequences in the perspective of developing appropriate tracking techniques [115]. In addition, ongoing studies on transthoracic high-frame-rate echocardiography could change the situation [104,106,116–119] in a near future. However, although the \textit{in vivo} feasibility of high-volume-rate 3-D echocardiography has been demonstrated by Provost \textit{et al.} [117], we are only in the very early stages of this promising imaging modality.

\textbf{In summary}—Volumetric speckle-tracking echocardiography is presently not feasible with the conventional ultrasound scanners. High-frame-rate 3-D echocardiography could potentially change the situation. Extensive technical and software developments, however, are still required.

\section*{26.10 Historical and clinical conclusion}

A complete echocardiographic examination must ideally allow quantification of the regional cardiac function, in particular for detecting regional myocardial anomalies under stress. Developed in the 1990s, the first way to quantify regional left ventricular wall motion was through endocardial (inner wall) tracking, a technique called “\textit{color kinesis}.” This technique enabled endocardium tracking by identifying pixel transitions from blood (cavity) to tissue (wall) on a frame-wise basis [120]. However, it soon became apparent that it was difficult to distinguish active from passive motion with this approach. \textit{Tissue Doppler imaging} thus emerged at the end of the 1990s (Figure 26.5). This new method was supposed to strongly improve physicians’ ability to detect regional dysfunctions through the measurement of the local deformations. But, as mentioned earlier, tissue Doppler imaging is angle dependent [40]. It was thus gradually replaced then supplanted by \textit{speckle tracking} (see Figure 26.5). The sequel has been discussed in this chapter: speckle-tracking echocardiography is now struggling to demonstrate its clinical utility, due to a lack of robustness and complexity of use. With time, the initially desired regional aspects have disappeared in favor of the global aspect. Fortunately, speckle-tracking echocardiography seems to find an honorable output as the GLS has been shown to be an early and reproducible marker of left ventricular dysfunction. We expect that the ongoing advances in medical ultrasound imaging will enable speckle tracking to return to the true source of its founding principles: detect regional myocardial dysfunctions.
References


[56] Gjesdal O, Helle-Valle T, Hopp E, et al. Noninvasive separation of large, medium, and small myocardial infarcts in survivors of reperfused...


