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iREVIEWS

STATE-OF-THE-ART PAPERS

Ultrafast Cardiac Ultrasound Imaging

Technical Principles, Applications, and Clinical Benefits

Maja Cikes, MD, PhD,*† Ling Tong, PhD,* George R. Sutherland, MD,* Jan D'hooge, PhD*

ABSTRACT

Several recent technical advances in cardiac ultrasound allow data to be acquired at a very high frame rate. Retrospective gating, plane/diverging wave imaging, and multiline transmit imaging all improve the temporal resolution of the conventional ultrasound system. The main drawback of such high frame rate data acquisition is that it typically has reduced image quality. However, for given clinical applications, the acquisition of temporally-resolved data might outweigh the reduction in image quality. It is the aim of this paper to provide an overview of the technical principles behind these new ultrasound imaging modalities, to review the current evidence of their potential clinical added value, and to forecast how they might influence daily clinical practice. (J Am Coll Cardiol Img 2014;7:812–23) © 2014 by the American College of Cardiology Foundation.

Cardiac ultrasound is the modality of choice for routine diagnostics given its low cost, high accessibility, and lack of ionizing radiation. Traditional 2-dimensional (2D) echocardiographic imaging is on the basis of a temporal resolution of ~25 to 40 Hz (i.e., data acquisition occurs every 25 to 40 ms), with later developments approximately doubling this. Although these frame rates are adequate to assess cardiac morphology and certain functional aspects, they do not allow the resolution of all cardiac mechanical events, as some of them are very short-lived. As such, potentially important diagnostic or prognostic information may get lost.

Recently, several technical advances have shown that 2D ultrasound imaging at a very high temporal resolution becomes possible through retrospective gating, plane wave/diverging wave imaging, and multiline transmit systems. These new imaging approaches grant insight into potentially new areas of myocardial mechanics and blood flow analysis. It is the aim of this paper to explain these developing technologies, review their potential benefits and current limitations, and examine how they might be applied to optimize cardiac diagnostics.

ULTRAFAST CARDIAC IMAGING: TECHNICAL CONSIDERATIONS

Ultrasound imaging is based on the acoustic pulse-echo measurement: an ultrasound pulse is transmitted and echo signals are subsequently received. Given the velocity of wave propagation in soft tissue (~1,530 m/s) and the propagation distance for a typical cardiac examination (~30 cm return-trip from the chest wall to the atrial roof), this measurement takes ~200 μs, implying that, for transthoracic cardiac applications, ~5,000 pulse-echo measurements can be made every second.

From the *Laboratory of Cardiovascular Imaging and Dynamics, Department of Cardiovascular Sciences, KU Leuven, Leuven, Belgium; and the †Department of Cardiovascular Diseases, University of Zagreb School of Medicine, University Hospital Center Zagreb, Zagreb, Croatia. This work was made possible through funding from the European Research Council under the European Union's Seventh Framework Program (FP7/2007-2013/ERC Grant Agreement number 281748) and the senior fellowship grant of the University of Leuven Research Council (KU Leuven-BOF). Dr. D'hooge has collaborative research projects with GE, Philips, and Epsilon Imaging. All other authors have reported that they have no relationships relevant to the contents of this paper to disclose. The first two authors contributed equally to this work.

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CONVENTIONAL ULTRASOUND IMAGING. To construct a 2D cardiac image, conventionally ~180 pulse-echo measurements are performed by sending pulses sequentially into 180 different directions within a plane covering a 90° angle (Fig. 1A). One image line is thus reconstructed for each ultrasound pulse transmitted. The construction of a single image thus takes 36 ms (180 × 200 μs), or ~28 frames can be created every second. To obtain optimal spatial resolution, the ultrasound beam is focused, which works best for large apertures. As such, the full aperture of the phased array transducer is used during transmit, resulting in the narrowest beam possible around the focal zone (Fig. 2A).

Given that the speed of sound and the dimensions of the heart are both fixed, the only way to increase frame rate is by lowering the number of transmit events required to construct a single frame. A straightforward way of achieving that is by limiting the field-of-view (i.e., narrowing the sector) or by reducing line density (i.e., increasing the spacing between adjacent firings), as both result in less transmit events/frame. The former approach preserves spatial resolution at the cost of the field-of-view, whereas the latter reduces spatial resolution, preserving the field-of-view. There thus exists an intrinsic relationship among field-of-view, spatial resolution, and temporal resolution.

CURRENT STATE-OF-THE-ART. To increase frame rate without compromising line density or sector width, a commonly-used technique is multiline acquisition (MLA) (1), which is implemented on most currently-available systems. In this approach, multiple neighboring lines are reconstructed simultaneously for each transmit beam (Fig. 1B). For example, in a 4MLA system, 4 lines will be reconstructed in parallel for each transmit beam, increasing the frame rate 4-fold. However, to enable this, the transmit beam needs to be broadened to ensure that the region covered by a given group of parallel receive lines is adequately insonified (Fig. 2B).

Broadening of the transmit beam is possible by using only part of the transducer's aperture during transmit (Fig. 2B). The smaller the aperture, the broader the beam, and the more lines can be reconstructed in parallel. However, reducing the aperture results in less energy being transmitted and, therefore, reduced image quality (i.e., signal-to-noise ratio [SNR]). Additionally, a broader beam will intrinsically result in reduced lateral resolution, which further degrades image quality. As a compromise, the number of parallel lines is therefore typically limited to 4.

RETROSPECTIVE GATING. A first approach to ultrafast imaging at preserved spatial resolution and field-of-view has been retrospective gating. In this approach, a large imaging sector is divided into several small subsectors. Each of these subsectors is imaged at a high frame rate, given its limited field-of-view, over 1 cardiac cycle. Subsequently, using retrospective electrocardiogram (ECG) gating, images of these subsectors are combined to obtain images of the full imaging sector at the original temporal resolution (2). However, ECG gating may fail if the ECGs derived during different cardiac cycles are markedly different. In such cases, motion matching is an alternative technique to combine subsector images (3). In particular, every 2 neighboring subsectors are slightly overlapping, and the local motion patterns extracted from these overlapping regions are temporally matched to exploit the periodicity of the cardiac motion to time-align 2 neighboring subsectors. In practice, a compromise has to be found

ABBREVIATIONS AND ACRONYMS

2D = 2-dimensional
ECG = electrocardiogram
MLA = multiline acquisition
MLT = multiline transmit
SNR = signal-to-noise ratio

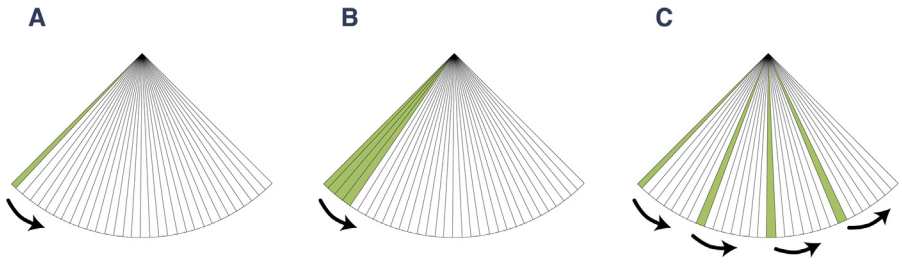

FIGURE 1 Ultrasound Scan Sequences

(A) Image scan scheme of conventional beam forming (single line acquisition); **(B)** multiline acquisition beam forming (e.g., 4 multiline acquisition); **(C)** multiline transmit beam forming (e.g., 4 multiline transmit). The small sectors indicate image lines required to form a full sector image; **green** indicates the region that needs to be covered by a transmit beam. The **arrows** indicate the direction of the scanning process.

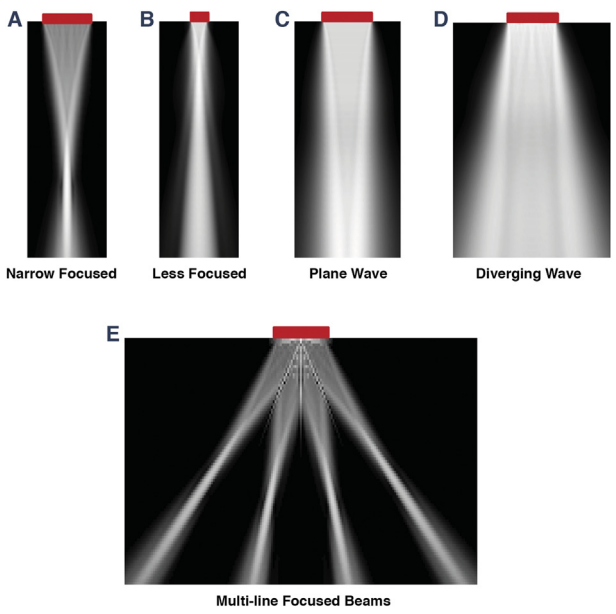

FIGURE 2 Different Kinds of Ultrasound Transmit Beams

(A) Narrow focused beam for single line acquisition/transmit; **(B)** less focused beam for 4 multiline acquisition (4MLA); **(C)** unfocused beam for massive multiline acquisition, plane wave; **(D)** diverging wave; and **(E)** narrow focused multiple transmit beams (e.g., 4MLA). The **red bar** on top of the beams indicates the transmit aperture size.

between the amount of subsectors to be combined and the frame rate obtained. This can be particularly challenging in patients with significant heart rate variability. Unfortunately, this approach is not applicable in atrial fibrillation.

PLANE-/DIVERGING-WAVE IMAGING. To avoid gating problems and prolonged acquisition times, plane- or diverging-wave imaging has been proposed. Originally, this imaging paradigm was implemented on linear array transducers (4–8), but more recently, it has proven applicable to phased arrays (9–13). In this technique, a plane/diverging wave is transmitted using the full aperture of the transducer (**Figs. 2C and 2D**). Such unfocused beams allow more lines to be reconstructed in parallel while keeping good energy penetration. In this way, some authors proposed to receive 16 lines in parallel for plane-wave transmits (9), whereas others proposed to receive all image lines at once by transmitting a very broad diverging wave (broader than the example given in **Fig. 2D**) (10). However, given the broad transmit beams, the spatial resolution degrades significantly. As a solution, coherent spatial compounding was proposed (4,5). In this technique, the same region is visualized from (slightly) different directions, and the final image is the average of all acquisitions (**Fig. 3**). In this way, the spatial resolution is significantly improved (**Fig. 4**). Although compounding in deep tissue is more difficult using a phased array due to its intrinsic small footprint, this has recently been demonstrated to be feasible (9). Obviously, compounding requires a greater number of transmits to scan the entire field-of-view, thereby reducing the effective gain in frame rate (**Fig. 4**). As a result, for (compounded) plane-wave imaging, the effective gain in frame rate is similar to that of a state-of-the-art 4MLA system (i.e., ~130 Hz for a 90° sector), resulting in very competitive performance of both systems (11). For diverging-wave imaging, the gain in frame rate can be much higher for a similar amount of compounding. In this way, a system operating at 1 to 2.5 kHz but at compromised spatial resolution and SNR was proposed (10). Although compounding over more transmit beams could theoretically solve this problem, the number of compounds should remain limited as motion between subsequent transmits will result in imperfect compounding with associated artifacts. Of note is that diverging-wave imaging makes harmonic imaging impractical or even impossible, as the transmitted acoustic energy is spread over a large region resulting in relatively small pressure amplitudes and, therefore, negligible generation of harmonics.

MULTILINE TRANSMIT IMAGING. As an alternative to parallel receive-beam forming, multiple focused beams can be transmitted simultaneously (i.e., multiline transmit [MLT]) (**Figs. 1C and 2E**). Although this approach was proposed more than 2 decades ago (14,15), it has received relatively little attention likely due to potential artifacts arising from cross-talk between simultaneously transmitted beams. Nevertheless, it was recently demonstrated that image quality of an MLT system could be very competitive to that of a conventional single-line acquisition system when set-up adequately (16,17). In contrast to the previously-mentioned approaches, the spatial resolution of an MLT image is not necessarily compromised (**Figs. 5A and 5B**), and harmonic imaging remains an option (18), both of which make this method particularly attractive. Moreover, MLT can easily be combined with MLA techniques (at the cost of spatial resolution) to gain an 8-fold increase in frame rate, which could probably be further improved to get a 12- to 16-fold increase without significant loss in image quality (17) (**Figs. 5C to 5F**).

APPLICATIONS OF ULTRAFAST CARDIAC IMAGING

B-MODE IMAGING. All of the previously-mentioned approaches have been demonstrated to enable

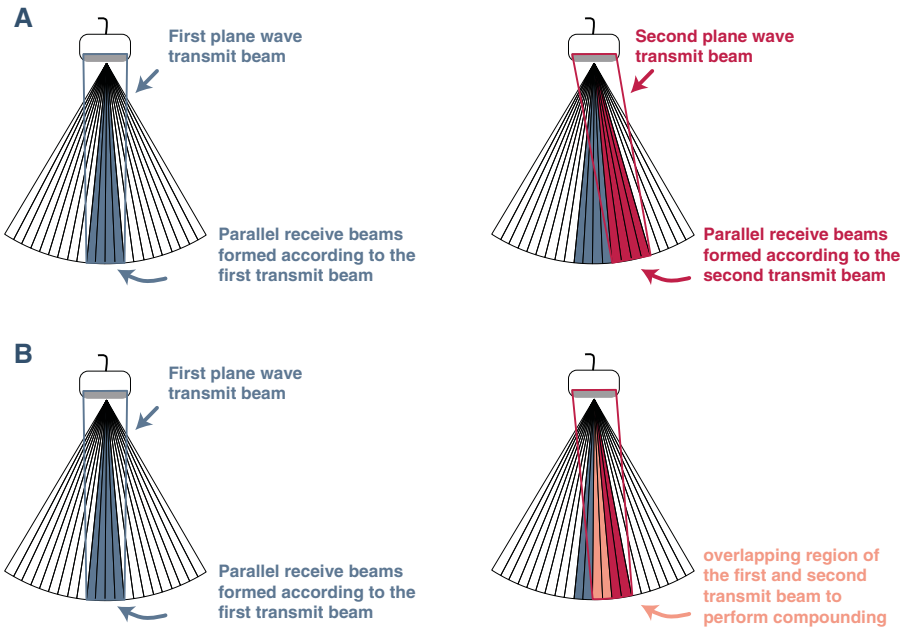

FIGURE 3 Plane Wave Imaging Scan Process Without and With Compounding

In this example, 4 parallel receive beams are formed for every transmit beam **(A)**. Without compounding, there is no overlap between the regions insonified by successive transmit beams. To perform compounding, successive transmit beams are partially overlapped **(B)**. The overlapped region is thus insonified from different directions. In this case, every image line within this overlapped region is imaged twice: once by the first and once by the second transmit beam. Adding receive beams at the same imaging angle but reconstructed from different transmit beams thus forms a compound image line at that angle. Obviously, more transmit beams are required to scan the whole field of interest when compounding is employed.

reconstruction of B-mode images. With retrospective ECG gating, images of a full 90° sector have been obtained at a frame rate of ~500 Hz (2,3). Compounded plane-/diverging-wave imaging was proven feasible in vivo, providing 90°-sector images at 316 Hz (9) or even up to 1 to 2.5 kHz (10). Very recently, similar results were reported using an MLT approach (17).

B-mode images can be reconstructed at high frame rates using these new imaging approaches; however, their added value for real-time display on the ultrasound system or for the quantification of cardiac morphology remains limited given the potential compromises in image quality and SNR. This, in itself, would therefore not warrant the implementation of these technologies.

MOTION AND DEFORMATION IMAGING. A first method for high frame rate acquisition to offer added value is by providing new insights into cardiac mechanics through enhanced motion/deformation analysis. Indeed, with current technology, regional myocardial motion/deformation can be quantified to some extent, but better resolution of short-lived mechanical events may offer new information on pathophysiological mechanisms.

Plane-wave imaging has been used to derive data on myocardial motion (12). To obtain the highest temporal resolution possible, compounding was avoided, thereby significantly compromising spatial resolution and SNR. However, in the setting of cardiac functional imaging, the gain in temporal resolution may outweigh the loss in spatial resolution/SNR. Indeed, regional velocity profiles similar to those obtained using state-of-the-art tissue Doppler imaging could be obtained using this approach (19,20). Similarly, diverging-wave imaging was recently demonstrated to provide physiologic motion profiles of the myocardium at high temporal (but reduced spatial) resolution (10). Although the added clinical value of obtaining myocardial motion/deformation information at high temporal resolution remains to be demonstrated, it is clear that obtaining this information becomes possible.

MECHANICAL ACTIVATION IMAGING. The imaging of myocardial motion/deformation at high temporal resolution has already demonstrated its clinical

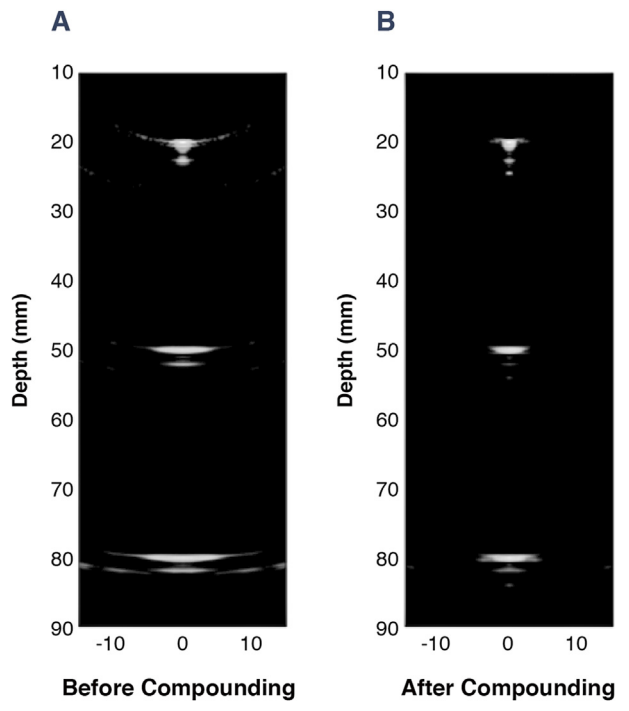

FIGURE 4 Simulated Images of 3 Point Targets Generated Using Diverging Wave Imaging

(A) Before coherent spatial compounding; **(B)** after coherent spatial compounding. The spatial resolution is improved after compounding. For clinical examples, please refer to Papadacci et al. (10).

potential through mechanical activation imaging. Myocardial mechanical activation is accompanied by brief (local) myocardial motion/deformation. Imaging the spatial distribution of the regional motion/deformation characteristics at high temporal resolution (<2 ms) can, therefore, provide information on the myocardial mechanical activation sequence. If electromechanical coupling is assumed to be constant throughout the ventricle, the mechanical activation wave could be representative for the preceding electrical depolarization sequence, with obvious applications in electrophysiology.

The concept of mechanical activation imaging was originally shown by increasing frame rate through coarse sampling (i.e., reducing line density) (21,22). Later on, it was demonstrated that retrospective gating could be used to better visualize and quantify this mechanical activation wave (2). As demonstrated in a study on open-chest dogs, motion matching also allows for adequate imaging of the mechanical activation wave (3). These authors also showed sensitivity to changes in electrical conduction resulting from myocardial ischemia, detecting regions with coronary flow obstruction >60% (3).

Despite these promising findings, mechanical activation imaging should preferably be performed within a single heartbeat. This was recently demonstrated to be feasible in a canine model using diverging-wave imaging (13) as well as in an open-chest pig model using plane-wave imaging (**Fig. 6**), which was recently carried out in our laboratory (23).

IMAGING VISCOELASTIC PROPERTIES OF THE MYOCARDIUM. High frame rate imaging could also provide insights into the intrinsic myocardial mechanical properties such as (local) stiffness or viscosity by assessing the characteristics of the propagation of shear waves through the myocardium. Shear waves complement compressional waves (acoustic waves) in that the direction of particle movement is transverse to the direction of wave propagation (**Fig. 7**). Shear waves propagate through tissue at a much lower velocity than acoustic waves, implying that the latter can be used to visualize shear wave propagation. Nevertheless, very fast imaging would be required (24). Interestingly, the propagation speed of shear waves and their frequency dependency are directly related to local viscoelastic properties. Assessing the characteristics of shear wave propagation thus allows for the estimation of viscoelastic properties of the medium through which they travel.

Shear waves can occur naturally inside of the human body (25). Alternatively, they can be generated in a more controlled manner by external vibrators (26) or by using the pushing force exerted by an ultrasound wave on its medium, that is, the acoustic radiation force (27). In the latter case, energy applied at the focus of the ultrasound beam will push the tissue in the direction of ultrasound wave propagation, resulting in generation of a shear wave that propagates transversely away from the “push” region. Although harmonic shear waves have been described and monitored, the 2D visualization of transient shear waves was first obtained in 2000 when fast imaging became available (28).

For cardiac applications, the propagation of shear waves was first characterized in the septum due to aortic valve closure (29). Here also, temporal resolution was appropriately increased by sparsely sampling the 2D image. In this way, an estimate of the myocardial viscoelastic properties could be made (29). More recently, acoustic radiation force-induced shear wave imaging based on plane-wave imaging on a linear array transducer was applied in a study of Langendorff-perfused isolated rat hearts (6). This study demonstrated the feasibility of measuring

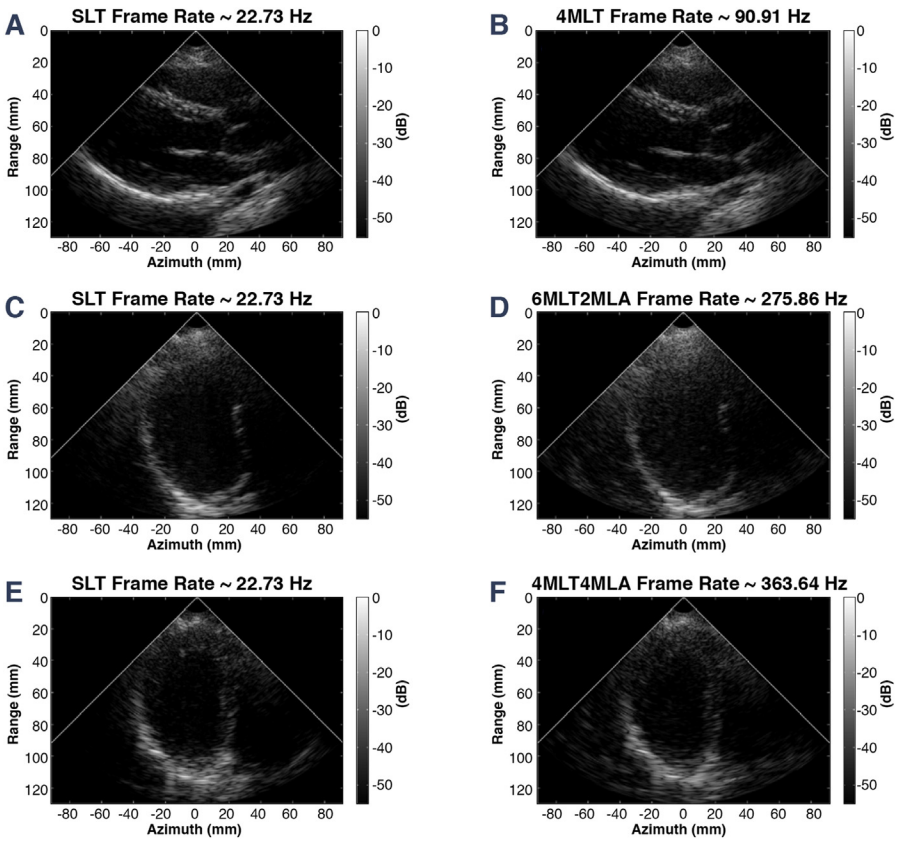

FIGURE 5 Comparison of Images Acquired Using the Conventional SLT Beam Forming and MLT Beam Forming

(A and B) Images acquired using single-line transmit (SLT) and 4 multiline transmit (MLT). **(C and D)** Images acquired using SLT and 6MLT-2 multiline acquisition (MLA). **(E and F)** Images acquired using SLT and 4MLT-4MLA.

myocardial stiffness locally as well as its dynamic behavior over the cardiac cycle with systolic stiffening of the myocardium (6). In addition, it was shown that peak systolic stiffness was relatively preload independent and clearly correlated with the myocardial inotropic state. More recently, the same group showed the technical feasibility of real-time shear wave imaging in vivo in an open-chest sheep model (30) and used this approach to determine local cardiac fiber orientation (31).

To date, cardiac shear wave imaging has remained mainly limited to the open-chest animal setting using linear array transducers. Only very recently has it been demonstrated in in vivo closed-chest animals and humans that shear wave imaging is feasible using phased arrays (32). However, some technical challenges remain. Moreover, generating a shear wave requires a push orthogonal to the myocardial wall, which can only be obtained—clinically—from a parasternal transducer position. True mapping of ventricular stiffness may thus remain impractical. Nevertheless, shear wave imaging could play an important future role in clinical cardiology.

BLOOD-POOL DOPPLER IMAGING. Despite its widespread clinical use, color flow mapping requires adapted imaging strategies that have a significant impact on temporal resolution. To limit these effects, in practice, the color box size is reduced to a relatively small region of interest.

As a solution, compound Doppler imaging has been proposed, which would allow the imaging of blood flow at a high frame rate (33). In this method, image acquisition is identical to compounded plane-wave imaging. Given the high frame rate of the resulting dataset, velocities can be estimated using established Doppler techniques. In this way, the acquisition for B-mode and Doppler imaging is identical, implying that blood flow can be estimated simultaneously at

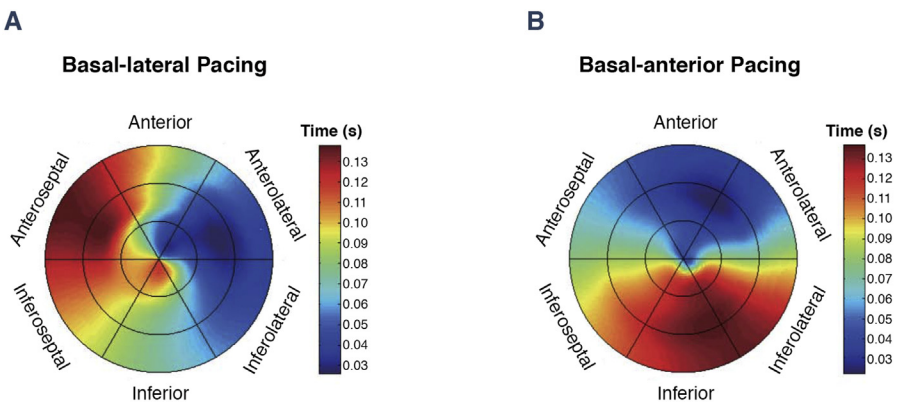

FIGURE 6 Example Cardiac LV Activation Maps Extracted From Datasets Acquired Using (Noncompounded) Plane Wave Imaging Based on an Open-Chest Pig Model

The left ventricle (LV) was under **(A)** basal-lateral pacing or **(B)** basal-anterior pacing. The frame rate was about 836 Hz.

each pixel in a wide 2D region of interest. Given the high sampling rate obtained in this manner, compound Doppler also provides the possibility to retrospectively obtain quantifiable Doppler spectra at all image points (24,33). Using a similar custom color flow approach, complex flow patterns in congenital heart disease were studied (8).

The previously-described Doppler approaches were implemented on linear-array transducers to facilitate fast imaging. Nevertheless, very recently, ultrafast Doppler imaging was also implemented on a phased-array transducer, using diverging waves to increase the field-of-view (34). This system was demonstrated to enable imaging of left ventricular blood flow patterns without the use of contrast agents at a temporal resolution of ~2,400 Hz (34). However, as only 2 compounded diverging waves were used to obtain the ultrahigh frame rate, spatial resolution was significantly compromised.

MYOCARDIAL PERFUSION IMAGING. The previously-mentioned gain in acquisition rate can also be used to generate flow data with a temporal resolution similar to what is currently used but at increased sensitivity to low flow rates. In this way, the capability of ultrafast Doppler to image intramyocardial blood flow was studied (35). Although these

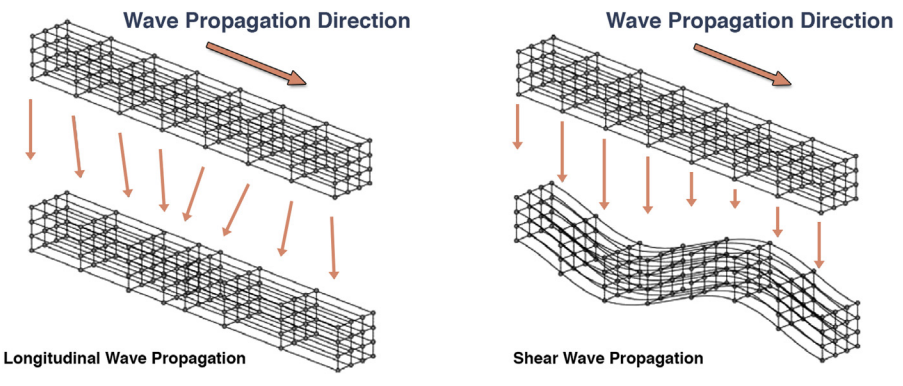

FIGURE 7 Types of Wave Propagation

The wave propagation is schematically presented by particles connected by massless springs that are displaced from their equilibrium position. **(Left)** Longitudinal wave (the movement of particles along the direction of wave propagation). **(Right)** Shear wave (the movement of particles transverse to the direction of wave propagation).

measurements were performed in an open-chest animal setup using linear-array transducers, disruptions in vascular flow during coronary occlusion could indeed be demonstrated.

Another modality of imaging myocardial perfusion is myocardial contrast echocardiography. Detection of signals generated by the microbubbles remains difficult, as they are embedded in echo signals from the surrounding myocardium. Ultrafast compounded plane-wave imaging has been employed in monitoring microbubble accumulation within the tissue vasculature with similar image quality and better contrast compared with conventional imaging (36).

An overview of published experimental and clinical studies focused on ultra-high frame rate imaging is given in **Table 1**.

POTENTIAL CLINICAL ROLE OF ULTRAFAST CARDIAC IMAGING

The periods of the heart cycle requiring the highest temporal resolution are the very short-lived intervals—the isovolumic contraction/relaxation periods. These are also the periods during which major differences between tissue Doppler imaging-based regional velocity and strain-rate curves occur; although relatively high velocities may be recorded, little change in regional strain-rate occurs (37). Although this might reflect only minor deformation changes during these time intervals, it might also be a matter of undersampling. Some of the events occurring during the isovolumic contraction period are the isovolumic acceleration and the septal flash (38,39). The former has been proposed as a measure of left ventricular contractility (38), whereas the latter was suggested as a marker of intraventricular dyssynchrony amenable to cardiac resynchronization therapy (39). Thus, further study of these short-lived intervals might provide more insight into myocardial contractility and the cardiac activation-contraction patterns, as well as new time-dependent parameters such as the rate of rise of systolic strain-rate. This information would obviously be valuable in the longitudinal follow-up of certain patient subpopulations, such as valvular disease, where it might provide additional data on optimal timing of valve replacement. However, reliable and robust quantification of any of these measures requires adequate temporal resolution of the underlying dataset.

Another field potentially benefiting from temporally-resolved myocardial deformation by ultrafast cardiac imaging is fetal echocardiography. The velocities and strain (-rates) of the fetal left and right ventricular walls can be readily obtained by tissue Doppler imaging (40,41). However, due to the high fetal heart rates, adequate temporal resolution becomes even more important. The optimal sampling requirements for fetal tissue velocity imaging were assessed at heart rates ~122 to 163 beats/min, where an acceptable 5% deviation corresponded to measurements performed above 150 to 200 Hz (42). More recently, the feasibility and reproducibility of speckle tracking-derived strain (-rate) analysis was tested in healthy and diseased fetuses, and showed relatively high interobserver and intraobserver discordance, with feasibility even more compromised with polyhydramnios (43). Having in mind the lower temporal resolution currently available in speckle tracking, it is most likely insufficient to resolve data, particularly the strain-rate. Thus, ultrafast cardiac imaging systems might likewise bring benefit to the field of fetal echocardiography.

Future technical advances might also provide adequate resolution of strain and strain-rate as derived from 2D speckle tracking as less angle-dependent measures of contractility. Furthermore, the acquisition of such high frame rate data on myocardial motion/deformation from multiple ventricular walls within the same heartbeat could be utilized in data fusion with electroanatomical mapping, thus providing more detailed and congruent/simultaneous information on the heart's *electromechanical properties*. This might improve the understanding of the heart's activation/contraction sequence, particularly in conduction abnormalities such as left bundle branch block and cardiac resynchronization therapy, as well as atrial arrhythmias.

Shear wave imaging should improve the understanding of the *viscoelastic properties* of the myocardium, which is of particular interest in heart failure, and especially cardiomyopathies. It might provide better insight into not only myocardial contractile function, but also its relaxation properties, potentially benefiting patient follow-up protocols and treatment guidance. Shear wave imaging might prove to become an additional modality of imaging for myocardial fibrosis and an adjunct in contractility assessment.

In addition to data on timing and quantification of myocardial motion/deformation and viscoelastic properties, ultrafast cardiac imaging can also provide better quantification of *complex intra-cardiac blood flows* (8), thus potentially optimizing hemodynamic assessment and 2D flow imaging, particularly in congenital heart disease. Furthermore, comprehensive intracardiac blood flow imaging might enable a better understanding of the pathophysiology underlying myocardial remodeling in various overload

TABLE 1 Overview of Published Experimental and Clinical Studies Focused on Ultra-High Frame Rate Imaging

First Author (Ref. #), Year	Acquisition Technique	Application	Study Setting	Imaging Target	Transducer; Frame Rate
Wang et al. (2), 2008	Retrospective ECG gating	B-mode imaging, mechanical activation imaging	Healthy volunteers	LV, abdominal aorta imaging at full-view frames, and axial displacement (STE); pulse wave propagation	Phased array; 481 Hz
Provost et al. (3), 2010	Motion matching	B-mode imaging, mechanical activation imaging	5 open-chest dogs in vivo	Conduction changes (electrical mapping) in ischemia	Phased array; 390–520 Hz
Couade et al. (7), 2010	Plane/diverging wave	Shear wave imaging	Healthy volunteer (carotid wall)	Arterial stiffness	Linear array; 8,000 Hz
Pernot et al. (6), 2011	Plane wave	Shear wave imaging	Langendorff-perfused isolated rat hearts	Myocardial stiffness (dynamic behavior over cardiac cycle)	Linear array; 12,000 Hz
Lovstakken et al. (8), 2011	Plane wave	Blood-pool Doppler imaging	Newborns, complex congenital heart disease	Complex 2D blood-flow patterns (STE approach)	Linear array Not reported
Hasegawa et al. (9), 2011	Compounded diverging (+plane) wave	B-mode imaging	Wire phantom, healthy volunteer	Feasibility of diverging wave using parallel beam forming	Phased array; 316 Hz at 90°
Provost et al. (13), 2011	Diverging wave	Mechanical activation imaging (motion and strain maps)	Open- and closed-chest in vivo dog model	Mechanical activation imaging in nonperiodic arrhythmias	Phased array; 2,000 Hz
Brekke et al. (20), 2011	Plane wave	3D B-mode imaging, 3D myocardial motion imaging (TDI)	5 healthy volunteers	Myocardial motion in 3D	Phased array; ≈500 Hz
Couade et al. (30), 2011	Plane wave	Real-time shear wave imaging	10 in vivo open-chest sheep	Local myocardial stiffness; myocardial elastic anisotropy	Linear array; 12,000 Hz
Bercoff et al. (33), 2011	Compounded plane wave	Ultrafast compound Doppler	Ultrasound phantoms, healthy volunteer (carotid artery)	Blood-pool Doppler imaging at HFR; simultaneous blood flow estimation at each pixel; quantifiable Doppler spectra at all image locations retrospectively	Linear array; 150 Hz (Doppler)
Tong et al. (12), 2012	Plane wave (noncompounded)	Motion imaging (color M-mode, velocity profiles); B-mode	Healthy volunteer	Feasibility of noncompounded plane-wave imaging in myocardial motion estimation	Phased array; 212 Hz
Cikes et al. (19), 2012	Plane wave	Motion imaging (TDI)	10 healthy volunteers	Plane wave feasibility in functional cardiac imaging	Phased array; 212 Hz
Lee et al. (31), 2012	Plane wave	Real-time shear wave imaging	5 in vitro porcine, 3 in vivo open-chest ovine hearts	Mapping local myocardial fiber orientation	Linear array; 8,000 Hz
Osmanski et al. (35), 2012	Compounded plane wave	Ultrafast Doppler imaging/myocardial perfusion imaging	5 in vivo open-chest sheep	Intramyocardial blood flow/ coronary blood flow dynamics (disruptions in vascular flow during coronary occlusion)	Linear array; 3,000 Hz
Couture et al. (36), 2012	Compounded plane wave	Myocardial perfusion imaging	Vessel phantom	Monitoring microbubble accumulation within the tissue vasculature	Linear array; 10.33 Hz
Song et al. (32), 2013	Diverging wave, pulse inversion harmonic imaging	Real-time shear wave imaging	Phantom, ex vivo pig heart, 7 healthy volunteers	Estimates of LV myocardium stiffness in end-diastole	Phased array; 3,850 Hz
Osmanski et al. (34), 2013	Diverging wave	Ultrafast Doppler imaging	Healthy volunteers	LV blood flow patterns during the heart cycle	Phased array; ~2,400 Hz
Papadacci et al. (10), 2014	Diverging wave	B-mode imaging, tissue velocity estimation	Phantoms, human in vivo	Cardiac imaging at very high frame rate with a large imaging sector	Phased array; 1,000–2,500 Hz
Tong et al. (17), 2014	Multiline transmit	B-mode imaging	Phantoms and healthy volunteers	Feasibility of multiline transmit beam forming for fast cardiac imaging	Phased array; 340–450 Hz
Prieur et al. (18), 2013	Multiline transmit	Second harmonic B-mode imaging	Wire phantom, in vivo LV	Feasibility of multiline transmit beam forming for second harmonic imaging	Phased array; not reported
Kanai et al. (21), 2001	Coarse sampling	Mechanical activation imaging	3 healthy volunteers, 2 aortic stenosis patients	Properties of myocardial vibrations —myocardial stiffness	Phased array; 600 Hz
Kanai (29), 2005	Coarse sampling	Shear wave imaging	5 healthy volunteers	Myocardial viscoelastic properties (Pulsive wave propagation)	Phased array; 450 Hz
Kanai (22), 2009	Coarse sampling	Mechanical activation imaging	Normal human heart	Propagation and dispersion properties of myocardial vibrations	Phased array; 500 Hz

2D = 2-dimensional; 3D = 3-dimensional; LV = left ventricle; STE = speckle tracking echocardiography; TDI = tissue Doppler imaging.

pathologies (congenital heart disease, valvular disease), ultimately providing better navigation for ventricular reconstructive surgery procedures. This might also be extended to imaging great vessel flows and defining aortic pathology and its treatment.

Ultrafast cardiac imaging may also impact *myocardial perfusion imaging* (35), as it can be used to increase the sensitivity of the Doppler system and, therefore, visualize low flow rates (as occurring in the smaller coronary arteries). Moreover, ultrafast imaging can increase the sensitivity of the ultrasound system to low contrast concentrations, thereby avoiding the need for higher-dose contrast injections and their associated image artifacts.

An overview of potential clinical applications of ultrafast cardiac imaging is given in Table 2.

TABLE 2 Potential Clinical Targets of Ultrafast Cardiac Imaging

Clinical Benefit	Ultrafast Cardiac Imaging Application
Motion and deformation during isovolumic periods	Plane wave velocity/deformation imaging Ultrahigh frame rate 3D TDI
Fetal echocardiography—diastolic events, isovolumic periods	Plane wave velocity/deformation imaging Ultrahigh frame rate 3D TDI
Resolution of strain (-rate) from 2D/3D STE	Plane wave velocity/deformation imaging Ultrahigh frame rate 3D TDI
Simultaneous motion/deformation data acquisition from multiple ventricular walls	Plane/diverging wave imaging
Single-cycle 3D volume acquisition	Plane/diverging wave imaging
Data fusion with electromechanical mapping; activation–contraction patterns	Retrospective gating
Myocardial stiffness/fibrosis/contractility	Shear wave imaging
Complex blood flow trajectories/dynamics	Ultrafast compound Doppler
Myocardial blood flow/perfusion imaging	Ultrafast compound Doppler Ultrafast signed power Doppler

Abbreviations as in Table 1.

POTENTIAL LIMITATIONS AND FUTURE DIRECTIONS IN ULTRAFAST CARDIAC IMAGING

One of the main drawbacks of high frame rate imaging is the trade-off with spatial resolution. The axial resolution—the ability of the ultrasound system to detect and distinguish echoes from structures at various depths along the beam axis, primarily depending on the length of the transmitted pulse—remains excellent. However, lateral resolution—the ability to detect and distinguish 2 closely-positioned echoes perpendicular to the beam axis, which is primarily dependent on the beam width—is often significantly impaired. Due to their large beam widths, plane- and diverging-wave imaging are examples of degraded lateral resolution. Nonetheless, MLT beam forming, with a demonstrated potential to overcome this drawback, utilizes focused transmit beams as in conventional beam forming, intrinsically preserving spatial resolution. Ultimately, all of the proposed fast imaging approaches are associated—some more than others—with a loss in SNR.

Harmonic imaging is very important in clinical practice to obtain good images in terms of spatial resolution and particularly SNR. It requires high acoustic pressures, which can be produced by increasing the transmit power and/or transmitting focused beams. For nongated high frame rate imaging methods, only MLT beam forming, in which focused beams are transmitted, has recently been shown to be suited for harmonic imaging (18). The feasibility of producing harmonics using plane waves or diverging waves remains a challenge due to their relatively low pressure amplitudes. In theory, it should be possible to generate harmonics in the near field by transmitting higher acoustic power when using plane waves or diverging waves. However, this may result in (thermal) safety issues, as well as a redefinition of current system hardware to allow for high power transmission. Nevertheless, good image quality is not a prerequisite for functional analysis of the heart, although it typically makes processing the data more robust.

Most of the high frame rate approaches have initially been performed using linear-array transducers, with later translation toward phased-array transducers. The reason for this is that linear arrays facilitate ultrafast imaging due to their intrinsic large field of view. Translating these methodologies to phased arrays brings along additional challenges. Nevertheless, solutions are being proposed, and phased-array fast ultrasound imaging has become a reality. Implementation of these imaging strategies on commercially-available clinical systems may, however, not be that straight-forward, as they not only require changes in imaging methodology, but they also bring challenges in data transfer rates and storage as well as changes in the system architectures. For instance, plane wave or diverging wave, allowing massive parallel MLA beams, would require a large number of hardware receive beamformers (≥16) or full software beam forming that could provide images in real-time. Both solutions would significantly increase the cost of the system, whereas the latter would also need to redefine the system architecture completely. For the MLT approach, a multilevel pulser would be required instead of the typical unipolar or bipolar pulsers used in cardiac ultrasound systems. Moreover, the transmitter should enable transmission of relatively long pulse trains (16). In addition, it is worthwhile to mention

that, in the MLT approach, the MLT beams are overlapping in the near field, which may cause (thermal) bioeffects. To avoid this problem, small delays could be introduced between adjacent MLT beams (44).

Although the current review has focused on fast 2D imaging, it is worth pointing out that the same imaging approaches are being used to speed up the acquisition process of 3-dimensional volume data where time resolution is even more of an issue, even with state-of-the-art volumetric imaging systems.

Finally, further studies are needed to test the necessity, accuracy, reproducibility, and reliability of the techniques before entering the area of clinical research, where the level of clinical evidence should further be validated (45).

CONCLUSIONS

Ultrafast cardiac imaging is a potential extension of echocardiographic imaging, with likely benefits. Provided that the technical advances allow the benefits to outweigh the drawbacks, ultrafast imaging might become an integral part of a new echocardiography toolbox, facilitating data analysis and providing additional insight into cardiac mechanics.

REPRINT REQUESTS AND CORRESPONDENCE: Prof. Jan D'hooge, Department of Cardiovascular Sciences, Medical Imaging Research Centre, University Hospital Gasthuisberg, Herestraat 49, 3000 Leuven, Belgium. E-mail: jan.dhooge@uzleuven.be.

REFERENCES

1. Shattuck DP, Weinshenker MD, Smith SW, von Ramm OT. Explososcan: a parallel processing technique for high speed ultrasound imaging with linear phased arrays. J Acoust Soc Amer 1984;75:1273–82.

2. Wang S, Lee WN, Provost J, Luo J, Konofagou EE. A composite high-frame-rate system for clinical cardiovascular imaging. IEEE Trans Ultrason Ferroelectr Freq Control 2008;55:2221–33.

3. Provost J, Lee WN, Fujikura K, Konofagou EE. Electromechanical wave imaging of normal and ischemic hearts in vivo. IEEE Trans Med Imaging 2010;29:625–35.

4. Tanter M, Bercoff J, Sandrin L, Fink M. Ultrafast compound imaging for 2-D motion vector estimation: application to transient elastography. IEEE Trans Ultrason Ferroelectr Freq Control 2002;49:1363–74.

5. Montaldo G, Tanter M, Bercoff J, Benech N, Fink M. Coherent plane-wave compounding for very high-frame rate ultrasonography and transient elastography. IEEE Trans Ultrason Ferroelectr Freq Control 2009;56:489–506.

6. Pernot M, Couade M, Mateo P, Crozatier B, Fischmeister R, Tanter M. Real-time assessment of myocardial contractility using shear wave imaging. J Am Coll Cardiol 2011;58:65–72.

7. Couade M, Pernot M, Tanter M, Prada C, Messas E, Fink M. Quantitative assessment of arterial wall biomechanical properties using shear wave imaging. Ultrasound Med Biol 2010;36:1662–76.

8. Lovstakken L, Nyrnes SA, Haugen BO, Torp H. Angle-independent quantification of complex flow patterns in congenital heart disease. Proceedings: IEEE International Ultrasonics Symposium 2011:1246–9.

9. Hasegawa H, Kanai H. High-frame-rate echocardiography using diverging transmit beams and parallel receive beam forming. J Med Ultrasonics 2011;38:129–40.

10. Papadacci C, Pernot M, Couade M, Fink M, Tanter M. High-contrast ultrasound imaging of the heart. IEEE Trans Ultrason Ferroelectr Freq Control 2014;2:288–301.

11. Tong L, Gao H, Choi HF, D'hooge J. Comparison of conventional parallel beam forming with plane wave imaging and diverging wave imaging for cardiac applications: a simulation study. IEEE Trans Ultrason Ferroelectr Freq Control 2012;59:1654–63.

12. Tong L, Hamilton J, Jasaityte R, Sutherland G, D'hooge J. Plane wave imaging for cardiac motion estimation at high temporal resolution: a feasibility study in-vivo. Proceedings: IEEE Ultrasonics Symposium 2012:228–31.

13. Provost J, Nguyen VT, Legrand D, et al. Electromechanical wave imaging for arrhythmias. Phys Med Biol 2011;56:L1–11.

14. Shirasaka T. Ultrasonic imaging apparatus. U.S. Patent 4.815.043, 1989.

15. Mallart R, Fink M. Improved imaging rate through simultaneous transmission of several ultrasound beams. Proc SPIE 1992;1733:120–30.

16. Tong L, Gao H, D'hooge J. Multi-transmit beam forming for fast cardiac imaging—a simulation study. IEEE Trans Ultrason Ferroelectr Freq Control 2013;60:1719–31.

17. Tong L, Ramalli A, Jasaityte R, Tortoli P, D'hooge J. Multi-transmit beam forming for fast cardiac imaging - experimental demonstration and in-vivo application. IEEE-Trans Med Imaging 2014;33:1205–19.

18. Prieur F, Denarie B, Austeng A, Torp H. Multi-line transmission in medical imaging using the second-harmonic signal. IEEE Trans Ultrason Ferroelectr Freq Control 2013;60:2682–92.

19. Cikes M, Tong L, Jasaityte R, Hamilton J, Sutherland G, D'hooge J. Can fast cardiac imaging using plane wave technology be used for cardiac motion estimation? A comparison to conventional tissue Doppler imaging in healthy volunteers. Eur Heart J Cardiovasc Imaging 2012;13 Suppl 1:i10 [abstr].

20. Brekke B, Torp H, Bjastad T, Støylen A, Aase SA. 3D tissue Doppler imaging with ultra high frame rate. Proceedings: IEEE Ultrasonics Symposium 2011:717–20.

21. Kanai H, Koiwa Y. Myocardial rapid velocity distribution. Ultrasound Med Biol 2001;27:481–98.

22. Kanai H. Propagation of vibration caused by electrical excitation in the normal human heart. Ultrasound Med Biol 2009;35:936–48.

23. Tong L. Novel Beam Forming Methods for Fast Cardiac Imaging Using Ultrasound. PhD Thesis. Leuven, Belgium: Leuven University Press, 2013.

24. Bercoff J. Ultrafast ultrasound imaging. In: Minin IV, Minin OV, editors. Ultrasound Imaging—Medical Applications. Rijeka, Croatia: InTech, 2011:3–24.

25. Kanai H, Yonechi S, Susukida I, Koiwa Y, Kamada H, Tanaka M. Onset of pulsatile waves in the heart walls at end-systole. Ultrasonics 2000;38:405–11.

26. Catheline S, Wu F, Fink M. A solution to diffraction biases in sonoelasticity: the acoustic impulse technique. J Acoust Soc Am 1999;105:2941–50.

27. Sarvazyan AP, Rudenko OV, Swanson SD, Fowlkes JB, Emelianov SY. Shear wave elasticity imaging: a new ultrasonic technology of medical diagnostics. Ultrasound Med Biol 1998;24:1419–35.

28. Sandrin L, Catheline S, Tanter M, Vinçonneau C, Fink M. 2D transient elastography. Acoust Imaging 2000;25:485–92.

29. Kanai H. Propagation of spontaneously actuated pulsive vibration in human heart wall and in vivo viscoelasticity estimation. IEEE Trans Ultrason Ferroelectr Freq Control 2005;52:1931–42.

30. Couade M, Pernot M, Messas E, et al. In vivo quantitative mapping of myocardium

stiffening and transmural anisotropy during the cardiac cycle. IEEE Trans Med Imaging 2011;30:295–305.

31. Lee W, Pernot M, Couade M, et al. Mapping myocardial fiber orientation using echocardiography-based shear wave imaging. IEEE Trans Med Imaging 2012;31:554–62.

32. Song P, Zhao H, Urban M, Manduca A, et al. Improved shear wave motion detection using pulse-inversion harmonic imaging with a phased array transducer. IEEE Trans Med Imaging 2013;32:2299–310.

33. Bercoff J, Montaldo G, Loupas T, et al. Ultrafast compound Doppler imaging: providing full blood flow characterization. IEEE Trans Ultrason Ferroelectr Freq Control 2011;58:134–47.

34. Osmanski BF, Pernot M, Fink M, Tanter M. In vivo transthoracic ultrafast Doppler imaging of left intraventricular blood flow pattern. Proceedings: IEEE International Ultrasonics Symposium 2013:1741–4.

35. Osmanski BF, Pernot M, Montaldo G, Bel A, Messas E, Tanter M. Ultrafast Doppler imaging of blood flow dynamics in the myocardium. IEEE Trans Med Imaging 2012;31:1661–8.

36. Couture O, Fink M, Tanter M. Ultrasound contrast plane wave imaging. IEEE Trans Ultrason Ferroelectr Freq Control 2012;59:2676–83.

37. Sutherland GR, Hatle L, Claus P, D'hooge J, Bijnens BH. Doppler Myocardial Imaging—A Textbook. Hasselt, Belgium: BSWK, 2006.

38. Vogel M, Cheung MM, Li J, et al. Noninvasive assessment of left ventricular force-frequency relationships using tissue Doppler-derived isovolumic acceleration: validation in an animal model. Circulation 2003;107:1647–52.

39. Parsai C, Bijnens B, Sutherland GR, et al. Toward understanding response to cardiac resynchronization therapy: left ventricular dyssynchrony is only one of multiple mechanisms. Eur Heart J 2009;30:940–9.

40. Harada K, Tsuda A, Orino T, Tanaka T, Takada G. Tissue Doppler imaging in the normal fetus. Int J Cardiol 1999;71:227–34.

41. Di Salvo G, Russo MG, Paladini D, et al. Quantification of regional left and right ventricular longitudinal function in 75 normal fetuses using ultrasound-based strain rate and strain imaging. Ultrasound Med Biol 2005;31:1159–62.

42. Elmstedt N, Lind B, Ferm-Widlund K, Westgren M, Brodin LÅ. Temporal frequency requirements for tissue velocity imaging of the fetal heart. Ultrasound Obstet Gynecol 2011;38:413–7.

43. Van Mieghem T, Giusca S, DeKonink P, et al. Prospective assessment of fetal cardiac function with speckle tracking in healthy fetuses and recipient fetuses of twin-to-twin transfusion syndrome. J Am Soc Echocardiogr 2010;23:301–8.

44. Madore B, White PJ, Thomenius K, Clement GT. Accelerated focused ultrasound imaging. IEEE Trans Ultrason Ferroelec Freq Contr 2009;56:2612–23.

45. Galderisi M, Henein MY, D'hooge J, et al., European Association of Echocardiography. Recommendations of the European Association of Echocardiography: how to use echo-Doppler in clinical trials: different modalities for different purposes. Eur J Echocardiogr 2011;12:339–53.

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