ISSN: 2249-7196
IJMRR/July 2023/ Volume 13/Issue 3/01-16
DR. K. SUDHAKAR/ International Journal of Management Research & Review

CNN-BASED ULTRASOUND IMAGE RECONSTRUCTION FOR ULTRAFAST DISPLACEMENT TRACKING

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ABSTRACT:

Ultrafast ultrasound imaging remains an active area of interest in the ultrasound community due to its ultrahigh frame rates. Recently, a wide variety of studies based on deep learning have sought to improve ultrafast ultrasound imaging. Most of these approaches have been performed on radio frequency (RF) signals. However, in-phase/quadrature (I/Q) digital beamformers are now widely used as low-cost strategies. In this work, we used complex convolutional neural networks for reconstruction of ultrasound images from I/Q signals. We recently described a convolutional neural network architecture called ID-Net, which exploited an inception layer designed for reconstruction of RF diverging-wave ultrasound images. In the present study, we derive the complex equivalent of this network; i.e., the Complex-valued Inception for Diverging-wave Network (CID-Net) that operates on I/Q data. We provide experimental evidence that CID-Net provides the same image quality as that obtained from RF-trained convolutional neural networks; i.e., using only three I/Q images, the CID-Net produces high-quality images that can compete with those obtained by coherently compounding 31 RF images. Moreover, we show that CIDNet outperforms the straightforward architecture that consists of processing the real and imaginary parts of the I/Q signal separately, which thereby indicates the importance of consistently processing the I/Q signals using a network that exploits the complex nature of such signals

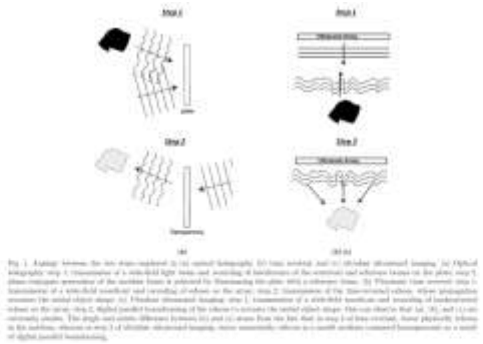
Index Terms: Biomedical imaging, deep learning, diffraction artifacts, displacement estimation, image reconstruction, speckle tracking, ultrafast ultrasound imaging.

INTRODUCTION U LTRAFAST ultrasound (US) imaging enables reconstructing full-view images from single acquisitions by insonifying the entire field of view at once, using unfocused transmit wavefronts such as plane waves (PWs) or diverging waves (DWs) [1]. Ultrasound images are then reconstructed from the received echo signals using the well-known delayand-sum (DAS) algorithm. Ultrafast US imaging thus breaks with the trade-off between field of view and frame rate inherent to conventional transmit-focused line-by-line scanning. This enables imaging large tissue regions at very high frame rates of multiple kilohertz, limited only by the round-trip propagation time of single acoustic waves. Imaging large tissue regions at such high frame rates is necessary for studying the most rapidly changing physical phenomena in the human body, such as tracking the propagation of naturally occurring or externally induced shear waves [2]– [6]. In the cardiovascular system, where a frame rate of several hundred hertz is needed for resolving tissue motion and flow patterns accurately [7]–[10], ultrafast imaging enables increased ensemble lengths, improving the robustness and sensitivity of displacement estimates significantly [10]. Several breakthrough US imaging modes based on motion estimation within a large field of view rely on ultrafast US imaging, such as shearwave elastography [2], ultrasensitive flow imaging [10], and functional US neuroimaging [11]. Because of the absence of transmit focusing, images obtained from ultrafast acquisitions are of low quality, suffering heavily from poor lateral resolution and low contrast [3]–[5], [8], [12], [13]. Both effects are related to the point spread function (PSF) of ultrafast US imaging systems, characterized by a broader main lobe (lower lateral resolution) and stronger diffraction artifacts (lower contrast) caused by side lobes (SLs), grating lobes (GLs), and edge waves (EWs), compared with conventional focused-US imaging systems. Naturally, low-quality images also limit the accuracy of subsequent displacement estimation methods involved in ultrafast US imaging modes [3], [5], [9]. The state-of-the-art solution for increasing the quality of ultrafast US imaging is coherent compounding, where a series of low-quality images, reconstructed from multiple, differently steered, unfocused wavefronts, are coherently summed [3], [12]. In [3], an image quality surpassing state-of-the-art multi-focus imaging was obtained by compounding 71 PW acquisitions, increasing the frame-rate by a factor of approximately seven. However, for analyzing motion at very high frame rates, coherent compounding suffers from two considerable disadvantages. Firstly, the increase in image quality is directly linked to the number of compounded acquisitions, which in turn is limited by the minimum frame rate necessary to analyze the underlying physical phenomenon of interest. Secondly, coherent compounding assumes, similarly to line-by-line scanning, that the region of interest is stationary for the duration of an acquisition sequence used to reconstruct a single frame. This assumption does not hold when imaging fast-moving tissue regions or complex flows, for which coherent compounding suffers from strong motion artifacts [13], [14]. The first issue is well exemplified in [3], in which Montaldo et al. demonstrated, in the context of shear-wave elastography, that the quality of estimated elasticity maps is directly linked to the number of compounded acquisitions, which in turn was limited to a maximum of twelve acquisitions to ensure a minimum frame rate of 1 kHz. In particular, displacement estimation in highly heterogeneous tissue regions, where the aforementioned diffraction artifacts were dominant, was a major obstacle. Issues due to diffraction artifacts hindering accurate displacement estimates have been reported for several methods, all of them suffering from the trade-off between image quality and frame rate [3], [5], [15]. The occurrence of severe motion artifacts when compounding multiple acquisitions of rapidly evolving physical phenomena (inter-frame displacement close to the effective wavelength) was discussed in [13], [14], [16], and motion compensation techniques were proposed to tackle this problem. They consist of estimating inter-acquisition displacement, using either conventional Doppler [14], [16] or 1-D correlation methods [13], and compensate for it before compounding all acquisitions to produce a motion-compensated high-quality image. However, these motion compensation techniques can also suffer from strong diffraction artifacts [13], as they are themselves based on displacement estimation from low-quality images, obtained from unfocused wavefronts. It thus remains unclear if such methods could help improve motion estimation in regions plagued by such artifacts. Consequently, there exists a great need for a robust displacement estimation technique that does not rely on multiple acquisitions to reconstruct consecutive frames. This is of particular interest in extreme conditions, when analyzing rapidly evolving physical phenomena in zones with highly heterogeneous echogenicities. In [17], we introduced a method for reconstructing highquality US images from single unfocused acquisitions. It consists of a backprojection-based DAS operation followed by the application of a convolutional neural network (CNN), specifically trained to reduce the diffraction artifacts inherent to the deployed ultrafast US imaging setup. Strong artifact reduction was demonstrated in simulated, in vitro, and in vivo environments.

The CNN-based image reconstruction method works strictly on a frame-by-frame basis and relies on the spatial information of each image only. Hence, it is completely agnostic to displacements that may occur between consecutive frames, making it a perfect fit for combination with state-ofthe-art image-based displacement estimation techniques. In a preliminary work [18] we showed that a CNN-based image reconstruction method may preserve the time-coherence of speckle patterns between consecutive frames, which is essential to any image-based displacement estimation technique. In this work, we propose an approach for estimating 2-D inter-frame displacements at maximum frame rates, by combining our CNN-based image reconstruction method [17] with a state-of-the-art 2-D speckle tracking algorithm. Although estimating the axial displacement (only) remains the standard in US imaging, 2-D displacement estimation is increasingly gaining attention in both flow and tissue motion applications [9], [19], [20], as it enables the analysis of more complex motion patterns. In elastography, 2-D displacement maps may be of interest to increase the quality and robustness of the estimated elasticity maps [21]. Also, 2-D speckle tracking represents an optimal fit for high-frame-rate displacement estimation since, unlike vector Doppler techniques, it does not rely on multi-angle acquisitions. Moreover, displacement estimation can be performed accurately from two consecutive frames only, whereas Doppler-based techniques usually require multiple consecutive frames to estimate the phase accurately. Since our aim is to tackle displacement estimation at maximum frame rates, the proposed approach relies only on single unfocused acquisitions to reconstruct consecutive frames and on two consecutive frames only to obtain 2-D displacement estimates. The primary goal of this work is to assess whether the diffraction artifact reduction and speckle restoration capabilities of our CNN-based image reconstruction method [17] can enable accurate estimation of displacements in zones initially shadowed by GL, SL, and EW artifacts

LITERATURE REVIEW

IN "M. TANTER AND M. FINK, "ULTRAFAST IMAGING IN BIOMEDICAL ULTRASOUND," IEEE TRANS. ULTRASON., FERROELECTR., FREQ. CONTROL, VOL. 61, NO. 1, PP. 102–119, JAN. 2014." Although the use of ultrasonic plane-wave transmissions rather than line-per-line focused beam transmissions has been long studied in research, clinical application of this technology was only recently made possible through developments in graphical processing unit (GPU)-based platforms. Far beyond a technological breakthrough, the use of plane or diverging wave transmissions enables attainment of ultrafast frame rates (typically faster than 1000 frames per second) over a large field of view. This concept has also inspired the emergence of completely novel imaging modes which are valuable for ultrasound-based screening, diagnosis, and therapeutic monitoring. In this review article, we present the basic principles and implementation of ultrafast imaging. In particular, present and future applications of ultrafast imaging in biomedical ultrasound are illustrated and discussed Conventional ultrasonography for medical ultrasound emerged in the 1970s, long after the invention of underwater sonar imaging by Paul Langevin at the beginning of the 20th century [1]. For this historical reason, the way in which we produce an image in medical ultrasound is intimately linked to the concept of echolocation in underwater acoustics. First, using single focused elements (A-mode), ultrasound devices were used to record backscattered echoes at depths along a focused beam line. During a short period in the 70s, ultrasonic images were then acquired by moving the focused transducer mechanically [2]. As a consequence, with the development of multiple element arrays and electronic focusing, line-per-line acquisition has become the core technology used in all ultrasound scanners [2]. In this way, medical ultrasound imaging is clearly a heritage of sonar. However, the concept of using plane waves to insonify a very large field of view in a single transmission and then building an image from the resulting backscattered echoes differs from sonar. In fact, this technique is more similar to optical concepts, particularly optical holography, which was invented by Denis Gabor in 1948 [3]. Indeed, optical holography is a method that allows a light field, which is generally the product of a light source scattered off a complete object, to be recorded on a holographic plate based on interference with a plane reference field. Using a nonlinear process, the interference of these waves transforms the phase information of the incident wave, which is necessary but unavailable, into experimentally available intensity information. The interference is later reconstructed when the original light field is no longer present, because of the absence of the original object. Illumination of the holographic plate by the complex conjugate of the plane reference wave allows for the generation of a complex conjugate of the incident wave (i.e., the hologram of the initial object) (Fig. 1). In ultrasound, such a holographic approach does not require nonlinear effects because reversible transducers can be used to record both the phase and amplitude of ultrasonic wavefields. A holographic experiment can be performed thanks to the concept of time reversal ultrasound, which was proposed by M. Fink in the 1980s [4], [5]. Similar to optical holography, time reversal ultrasound involves transmission of a wide field-of-view wavefront into the medium, which is followed by recording of backscattered echoes on an ultrasonic array. Time reversing of these echoes, which consists of re-emitting previously recorded echoes in a reversed chronology, builds a wavefield that refocuses optimally on the initial object during backpropagation. At the time of refocusing, the spatial distribution of the wavefield corresponds to the physical image of the object. Such time-reversed propagation can be used to either physically refocus on selected targets for destruction/aberration corrections in adaptive focusing [6] or to virtually recreate an image of the initial scattering object in the computer (Fig. 1). This second approach, corresponding to a numerical time reversal experiment, in which the time-reversed echoes are refocused on the initial object in a numerical model of the propagation medium (with a constant and homogeneous sound speed), was introduced in the 1980s independently of the time-reversal concepts. It is usually referred to as digital parallel receive beamforming in the literature. Therefore, ultrafast ultrasound imaging utilizes concepts that come from optical holography and ultrasonic time reversal. Overall, this imaging technique consists of transmitting a wide field-of-view beam into the medium, recording the resulting backscattered echoes, and finally performing digital parallel beamforming (or time-reversal focusing) of the echoes to computationally build the final ultrasonic image from a single transmission. Optical holograms are recorded using a single flash of light that illuminates a complete scene and is imprinted on a recording medium. Subsequently, a second illumination is utilized to recreate the phase conjugate of the initial interference. In contrast, ultrafast ultrasound insonifies the medium with a single plane wave, and the backscattered echoes are time reversed (wideband analog of phase conjugation) and numerically backpropagated to recreate the initial ultrasonic scene. More than sixty years after their discovery, optical holograms are mainly used in art. However, this technology has the potential to be utilized for engineering next generation storage devices, because it displays more storage capacity than Blu-Ray platforms (i.e., the entire volume of the recording media can be exploited, instead of just the surface). In ultrasound, the development of ultrafast scanners based on plane-wave imaging and parallel beamforming processing have the potential to revolutionize medical ultrasound imaging; however, with time this technology could also be used for advancements in other fields.



IN "J. BERCOFF, M. TANTER, AND M. FINK, "SUPERSONIC SHEAR IMAGING: A NEW TECHNIQUE FOR SOFT TISSUE ELASTICITY MAPPING," IEEE TRANS. ULTRASON., FERROELECTR., FREQ. CONTROL, VOL. 51, NO. 4, PP. 396–409, APR. 2004" —Supersonic shear imaging (SSI) is a new ultrasound-based technique for real-time visualization of soft tissue viscoelastic properties. Using ultrasonic focused beams, it is possible to remotely generate mechanical vibration sources radiating low-frequency, shear waves inside tissues. Relying on this concept, SSI proposes to create such a source and make it move at a supersonic speed. In analogy with the "sonic boom" created by a supersonic aircraft, the resulting shear waves will interfere constructively along a Mach cone, creating two intense plane shear waves. These waves propagate through the medium and are progressively distorted by tissue heterogeneities. An ultrafast scanner prototype is able to both generate this supersonic source and image (5000 frames/s) the propagation of the resulting shear waves. Using inversion algorithms, the shear elasticity of medium can be mapped quantitatively from this propagation movie. The SSI enables tissue elasticity mapping in less than 20 ms, even in strongly viscous medium like breast. Modalities such as shear compounding are implementable by tilting shear waves in different directions and improving the elasticity estimation. Results validating SSI in heterogeneous phantoms are presented. The first in vivo investigations made on healthy volunteers emphasize the potential clinical applicability of SSI for breast cancer detection. By inducing motion in soft tissue, elastography has become a wide medical application field to study soft tissue mechanical properties. Several techniques arose according to the type of mechanical excitation chosen (static compression, monochromatic, or transient vibration) and the way these excitations are generated (externally or internally). Different imaging modalities can be used to estimate the resulting tissue displacements: ultrasound or magnetic resonance (MR) imaging. Static elastography [1], a pioneer in this field, is limited by various artifacts due to unknown boundary conditions. The MR elastography [2] gives very good resolution, but the acquisition time (~20 minutes) limits the application to static organs. A few years ago, our group introduced transient elastography in order to overcome these limitations and to provide quantitative elasticity maps of tissues [3]. Based on real time imaging of transient shear waves generated in the body by an external mechanical vibrator, this technique has the advantage of being insensitive to patient motion and boundary condition artifacts and gave promising re sults on in vivo breast cancer detection. However, the clinical applicability of this technique was limited by the use of heavy and bulky external vibrators that vibrate with a unique shear spatial directivity pattern. As a consequence, biased elasticity estimation could occur if some parts of the image are not reached by the shear wave propagation. An alternative solution to external vibrations is to use the acoustic radiation force created by an ultrasonic focused beam to investigate tissues mechanical properties. Several groups are currently investigating this approach. Fatemi and Greenleaf [4] proposed, with the vibroacoustography technique, the combination of two different ultrasonic beams with slightly different center frequencies to generate a low-frequency beat mode that can be detected with an external microphone. Another approach, investigated by Sarvazyan et al. [5], shear wave elasticity imaging (SWEI) and Nightingale et al. [6], acoustic radiation force impulse (ARFI) imaging consists in focusing an ultrasonic beam deep in tissues during a long time (typically 100 µs) and measuring the resulting displacement at focus using ultrasonic correlation-based techniques. The displacement induced at focus as a function of time can be linked to the local viscoelastic properties of the tissue. However, these mechanical displacements depend on several parameters, such as the acoustic beam geometry, the ultrasonic absorption coefficient, and the shear wave heterogeneities in the focal spot, making these techniques unable to provide quantitative estimation of tissue viscoelastic parameters. In this paper we report a new technique, called supersonic shear imaging (SSI), providing quantitative shear modulus mapping of an organ in less than 30 ms. Combining the advantages of the different approaches presented above, SSI relies on the acoustic radiation force to remotely generate low-frequency shear waves in tissues and can be achieved using the same piezoelectric arrays as the ones used in conventional ultrasonic scanners. This radiation force acts as a dipolar source of shear waves and mainly radiates in transverse directions. We propose, with SSI, to create quasiplane shear waves of stronger amplitude by moving the shear source at a supersonic speed. Such a shear source, which moves faster than the shear waves, can be created by successively focusing the ultrasonic "pushing" beam at different depths. All resulting shear waves interfere constructively along a Mach cone, creating two quasiplane shear wave fronts propagating in opposite directions. The angle between the two plane waves is proportional to the ratio between the shear-wave speed and the speed of the moving source, i.e., to the Mach number. The ultrafast, ultrasonic scanner developed for the technique is able to generate this supersonic shear source and image the propagation of the resulting transient plane shear waves by reaching frame rates of a few kilohertz. Shear modulus can be estimated by imaging the shear-wave propagation in a source-free region. This approach, which differs from other radiation force techniques that try to extract viscoelastic properties from displacements at focus, has the advantage to be quantitative and realizable in a few milliseconds. Furthermore, establishing such a supersonic regime brings several essential innovations. First, constructive interference between shear waves creates a cumulative effect that induces high-mechanical displacements in the medium (up to 100 µm in phantoms and 40 µm in vivo). This is an indispensable condition for the in vivo feasibility of the technique, particularly in strongly viscous media (breast, liver). Second, the supersonic regime generates two spatially extended plane shear waves, increasing the area in which mechanical shear information is available. Third, changing the speed of the moving "pushing" beam allows us to change the Mach cone angle, then insonify the same medium with different steered plane waves. With this method, called shear compounding, it is possible to gather the same mechanical information from different angles of view and, therefore, to improve the robustness of the technique

IN "G. MONTALDO, M. TANTER, J. BERCOFF, N. BENECH, AND M. FINK, "COHERENT PLANE-WAVE COMPOUNDING FOR VERY HIGH FRAME RATE ULTRASONOGRAPHY AND TRANSIENT ELASTOGRAPHY," IEEE TRANS. ULTRASON., FERROELECTR., FREQ. CONTROL, VOL. 56, NO. 3, PP. 489–506, MAR. 2009" —The emergence of ultrafast frame rates in ultrasonic imaging has been recently made possible by the development of new imaging modalities such as transient elastography. Data acquisition rates reaching more than thousands of images per second enable the real-time visualization of shear mechanical waves propagating in biological tissues, which convey information about local viscoelastic properties of tissues. The first proposed approach for reaching such ultrafast frame rates consists of transmitting plane waves into the medium. However, because the beamforming process is then restricted to the receive mode, the echographic images obtained in the ultrafast mode suffer from a low quality in terms of resolution and contrast and affect the robustness of the transient elastography mode. It is here proposed to improve the beamforming process by using a coherent recombination of compounded plane-wave transmissions to recover high-quality echographic images without degrading the high frame rate capabilities. A theoretical model is derived for the comparison between the proposed method and the conventional B-mode imaging in terms of contrast, signal-to-noise ratio, and resolution. Our model predicts that a significantly smaller number of insonifications, 10 times lower, is sufficient to reach an image quality comparable to conventional B-mode. Theoretical predictions are confirmed by in vitro experiments performed in tissue-mimicking phantoms. Such results raise the appeal of coherent compounds for use with standard imaging modes such as B-mode or color flow. Moreover, in the context of transient elastography, ultrafast frame rates can be preserved while increasing the image quality compared with flat insonifications. Improvements on the transient elastography mode are presented and discussed. Ultrasound imaging has become a major medical imaging modality in the last 30 years. The main technological advance that led to its proliferation was the early development of real-time imaging capabilities. The data acquisition rate in medical ultrasonic imaging devices is only limited by the acoustic propagation velocity in the tissues. Typically, in current systems, the number of transmit events is equal to the number of scan lines to be formed, which limits the frame rate to about 30 to 40 frames per second (fps). However, increasing the frame rate in ultrasound imaging will pave the way to tremendous new applications of medical ultrasound. It will lead to future real-time 3-D imaging systems or to improved tracking of the movements of the heart during the cardiac cycle. It will also enable the visualization of transient events such as shear mechanical wave propagation for elasticity imaging or even indirect imaging of the mechanical effects of electric action potentials. High frame rate will also be used for image enhancement methods such as video integration or compounding imaging approaches. Although its capabilities have only recently been fully developed, the concept of ultrafast echographic imaging is not new. In 1979, Delannoy et al. [1], [2] proposed use of parallel processing approaches to produce entire frames simultaneously from a single acoustic pulse. Their scanner was reaching a frame rate of 1000 images per second with 70 lines per frame. In 1984, Shattuck et al. implemented a parallel processing approach for a phased-array sector scanner enabling the simultaneous acquisition of several B-mode lines from each transmitted acoustic burst [3]. This approach was called “explososcan” by the authors, and the proof of concept was validated in vivo. One of the first explososcan systems was designed based on the single transmission of a “fat” ultrasonic beam and the parallel processing of 4 ultrasonic beams in the receive mode [4], [5]. This led to a system relying on a data acquisition rated increased by a factor of 4. Beyond this first successful attempt, they envisioned that this method could be, at least conceptually, extended to the imaging of a complete tomographic plane from the echoes produced by a single transmitted pulse, provided that this pulse fully illuminated the region of interest. More than 15 years later, Fink and co-authors [6]–[8] applied successfully the concept of plane wave illuminations leading to ultrafast frame rates higher than 5000 fps. The goal was to image in real time the transient propagation of shear mechanical waves in human tissues for the assessment of local viscoelastic properties, a technique dubbed “transient elastography.” This ultrafast approach led to the first experimental in-vivo investigation for breast cancer diagnosis [9]. Finally in 2004, by combining this ultrafast imaging modality with a remote palpation induced by the radiation force of ultrasonic focused beams, they provided an efficient way to build viscoelasticity maps of human tissues using a conventional echographic probe, the so-called supersonic shear imaging (SSI) technique [10], [11]. Another approach for reaching a high frame rate that used nondiffracting beams was proposed by Lu and Greenleaf in the early 1990s [12]–[14]. Following these pioneer works, Lu developed a theory in which a pulsed plane wave was used in transmission, and limited-diffraction array beam weighting was applied in reception to produce a spatial Fourier transform of the object function for 3-D image reconstruction. Later, he proposed the use of spatial compounding with different steered plane waves or limited diffraction beams added in a coherent way (enhancing resolution) or added in an incoherent way (reducing speckle) [15]–[17]. Ultrasound compound imaging for medical imaging dates back to the early 1980s [18] and has been widely studied in the last decade [19], [20]. To date, this concept of compound imaging is mainly associated with an incoherent addition of several image frames in an attempt to cancel out random variations (noise) and hence increase the signal to noise ratio [20], [21]. The reported benefits of image compounding in ultrasonography include reduced speckle, reduction in artifacts (clutter, shadowing, echo drop-out), higher contrast, and better visibility of lesion margins (making lesions more conspicuous). The "incoherent" terminology is related to the summation method that acts on the intensity images to smooth the speckle noise. Contrary to incoherent compound imaging, coherent wave compounding has not been extensively studied. It consists of the recombination of backscattered echoes from different illuminations achieved on the acoustic pressure field (as opposed to the acoustic intensity for the incoherent case). Although the concept has already been proposed [22] and mentioned in the context of limited diffraction beams [17], to date, no extensive work has been done to study its performance and advantages. We propose in this paper a theoretical model quantifying the performance of the coherent plane-wave-compounding approach. The interesting finding reported here is that the number of insonifications needed to achieve a given image quality is significantly lower than for classical approaches to B-mode. This characteristic opens new potential applications to the method such as transient elastography and fast color-flow imaging. Coherent plane-wave compounding of images obtained with different angles presents strong conceptual analogies with the synthetic aperture method [23], [24]. In the synthetic aperture approach, the ultrasonic array is fired element by element, and the complete set of impulse responses between each transmit-and-receive element is recorded. It is then possible to post-process these data to generate a synthetic image relying on both transmit-andreceive focusing for each pixel of the image. It has been intensely discussed in the literature whether synthetic imaging could give better images than conventional B-mode images, and how they will be affected by tissue motion and limited signal-to-noise ratio. A fundamental problem in synthetic aperture imaging is the poor signal-to-noise ratio in the images, because a single element is used for emission. This gives much lower emitted energy compared with using the full aperture in conventional imaging and, therefore, limits the depth of penetration. Coherent plane-wave imaging solves at least partially these limitations of synthetic aperture imaging. First, the transmission of a plane wave on the complete array generates a much higher wavefield than in the synthetic aperture approach. Second, as it will be shown in this paper, the reconstruction of high-quality echographic images in a small amount of time enables measurement of tissue displacement or flow-imaging processing. As already mentioned, in the SSI, the ultrafast frame rate is reached by limiting the illumination to a single plane-wave transmission and storing the backscattered signals in RAM memory. The image formation is then reduced to a conventional beamforming approach in the receive mode. Consequently, a single insonification is sufficient to produce an echographic image. The time between 2

ISSN: 2249-7196
IJMRR/July 2023/ Volume 13/Issue 3/01-16
DR. K. SUDHAKAR/ International Journal of Management Research & Review
consecutive images is only limited by the propagation time in tissues. As an example, the acquisition of 50 mm depth images can be theoretically produced at a frame rate of ~10 Kf/s. Due to the lack of transmit focusing, the echographic image quality is of course degraded in terms of contrast and resolution, and such degradation has consequences for the robustness of the elastography mode. In a previous work [25], we demonstrated the interest of incoherent plane-wave compounding at ultrafast frame rates for the estimation of transverse tissue displacements in transient elastography. In that case, the goal was to improve the estimation of lateral motion by increasing the number of independent speckle noise patterns used for the speckle tracking estimation. As speckle patterns recorded for different illumination angles were decorrelated, the variance of lateral motion estimates was improved following a square root dependence versus the number of angles. In this article, we propose investigation of the concept of coherent plane-wave compounds for improving echographic image quality without resorting to very high frame rate capabilities. The new imaging sequence is based on the coherent summation of ultrafast images obtained from plane wave transmissions at different angles.

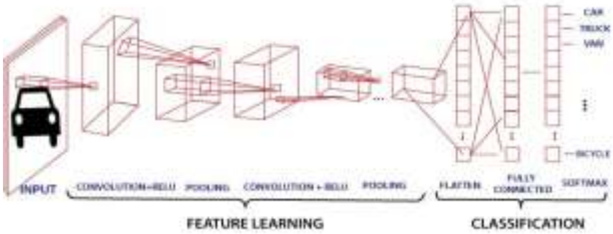
IN "P. SANTOS ET AL., "NATURAL SHEAR WAVE IMAGING IN THE HUMAN HEART: NORMAL VALUES, FEASIBILITY, AND REPRODUCIBILITY," IEEE TRANS. ULTRASON., FERROELECTR., FREQ. CONTROL, VOL. 66, NO. 3, PP. 442–452, MAR. 2019" Left ventricular myocardial stiffness could offer superior quantification of cardiac systolic and diastolic function when compared to the current diagnostic tools. Shear wave elastography in combination with acoustic radiation force has been widely proposed to non-invasively assess tissue stiffness. Interestingly, shear waves can also result from intrinsic cardiac mechanical events (e.g., closure of valves) without the need for external excitation. However, it remains unknown whether these natural shear waves always occur, how reproducible they can be detected and what the normal range of shear wave propagation speed is. The present study therefore aimed at establishing the feasibility of detecting shear waves created after mitral valve closure (MVC) and aortic valve closure (AVC), the variability of the measurements, and at reporting the normal values of propagation velocity. Hereto, a group of 30 healthy volunteers was scanned with high frame rate imaging (>1000 Hz) using an experimental ultrasound system transmitting a diverging wave sequence. Tissue Doppler velocity and acceleration were used to create septal color M-modes, on which the shear waves were tracked and their velocities measured. Overall the methodology was capable of detecting the transient vibrations that spread throughout the intraventricular septum in response to the closure of the cardiac valves in 92% of the recordings. Reference velocities of 3.2±0.6 m/s at MVC and 3.5±0.6 m/s at AVC were obtained. Moreover, in order to show the diagnostic potential of this approach, 2 patients (one with cardiac amyloidosis and one undergoing a dobutamine stress echocardiography) were scanned with the same protocol and showed markedly higher propagation speeds: the former presented velocities of 6.6 m/s and 5.6 m/s; the latter revealed normal propagation velocities at baseline, and largely increased during the dobutamine infusion (>15 m/s). Both cases showed values consistent with the expected changes in stiffness and cardiac loading conditions. For the quantification of both cardiac systolic and diastolic function, clinicians largely rely on the interpretation of changes in chamber geometry during the cardiac cycle and velocity measurements of the blood flow. Left ventricular (LV) myocardial stiffness, on the other hand, provides information on the forces acting inside the myocardium, thus allowing a more direct assessment of its systolic and diastolic properties [1], [2]. However, myocardial stiffness measurements during contraction and relaxation are difficult to obtain non-invasively and limited human data is available. Different shear wave elastography implementations have been presented in order to non-invasively assess myocardial stiffness with ultrasound. All approaches capture the tissue's transient response to a local perturbation in order to infer certain mechanical properties of the myocardium (e.g., stiffness and viscosity). However, they fundamentally differ on the excitation source, which can be either exogenous (mechanical [3] or ultrasonic [4]–[8]) or endogenous (closing of cardiac valves [9]–[14]). Because the latter occur naturally – e.g., after mitral valve closure (MVC) and aortic valve closure (AVC) – their generation does not require additional mechanical vibrators nor special transducers/excitations, hence often being referred to as "natural" or "passive" shear waves. Compared to the most commonly used acoustic radiation force excitation, natural shear waves are also characterized by higher displacements (∼100 µm [13] vs. ∼1 µm [5]), potentially allowing them to travel and be tracked over larger distances. Given the dimensions of the heart and the typical shear wave propagation speed of 1 to 10 m/s, cardiac shear waves are short-lived. As an example, a wave propagating at 5 m/s will only take 16 ms to travel across a 8 cm-long septum. Hence, their detection using ultrasound only became possible with high frame rate (HFR) imaging [15], [16], initially by using sparse sector scanning [17] and subsequently by taking advantage of ECG-gating techniques [10] or anatomical sparse scanning [11] to improve the spatial sampling of these patterns. The development of diverging wave (DW) imaging for phased arrays [18]–[20] then provided full-sector cardiac images from as few as one transmission (i.e., with frame rates only limited by the acoustic time-of-flight). DW imaging has been used to jointly study MVC and AVC waves in animal models [13], although the necessary wide defocused transmissions lead to a loss in lateral resolution and signal-to-noise ratio (SNR). These disadvantages can partially be overcome by coherent compounding [21], which could thus improve the visualization and analysis of those events. Although transient vibrations at end-diastole and endsystole have previously been described, measuring MVC and AVC shear waves jointly has been limited to animal models or small human populations. As such, it remains of clinical relevance to understand whether these waves can be detected systematically, what their normal range of propagation velocities is and how reproducible these velocities can be assessed [22]. Moreover, contrasting the results presented in acoustic radiation force studies with the ones obtained for naturally occurring shear waves is of interest given the difference in both the excitation profile and timing at which the shear waves are evaluated. The present study thus aimed at assessing the feasibility of measuring the velocity of the shear waves created by both MVC and AVC in a single recording taken in a clinical setting. A population of healthy subjects was scanned and their MVC and AVC wave velocities were computed from HFR imaging data. To demonstrate the potential of the technique to distinguish between normal and abnormal myocardial states during the systolic and diastolic phases of the cardiac cycle, AVC and MVC shear wave velocities were also measured in a patient with amyloidosis and a patient undergoing dobutamine stress echocardiography.

ALGORITHM

Convolutional Neural Network is one of the main categories to do image classification and image recognition in neural networks. Scene labeling, objects detections, and face recognition, etc., are some of the areas where convolutional neural networks are widely used.

CNN takes an image as input, which is classified and process under a certain category such as dog, cat, lion, tiger, etc. The computer sees an image as an array of pixels and depends on the resolution of the image. Based on image resolution, it will see as **h * w * d**, where h= height w= width and d= dimension. For example, An RGB image is **6 * 6 * 3** array of the matrix, and the grayscale image is **4 * 4 * 1** array of the matrix.

In CNN, each input image will pass through a sequence of convolution layers along with pooling, fully connected layers, filters (Also known as kernels). After that, we will apply the Soft-max function to classify an object with probabilistic values 0 and 1.



Convolution Layer

Convolution layer is the first layer to extract features from an input image. By learning image features using a small square of input data, the convolutional layer preserves the relationship between pixels. It is a mathematical operation which takes two inputs such as image matrix and a kernel or filter.

- The dimension of the image matrix is **h×w×d**.
- The dimension of the filter is **f_h×f_w×d**.
- The dimension of the output is **(h-f_h+1)×(w-f_w+1)×1**.

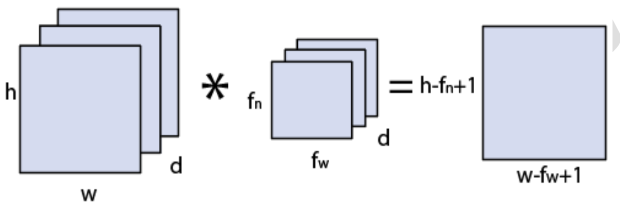


Image matrix multiplies kernl or filter matrix

Let's start with consideration a 5*5 image whose pixel values are 0, 1, and filter matrix 3*3 as:

$$\begin{bmatrix} 1 & 1 & 1 & 0 & 0 \\ 0 & 1 & 1 & 1 & 0 \\ 0 & 0 & 1 & 1 & 1 \\ 0 & 0 & 1 & 1 & 0 \\ 0 & 1 & 1 & 0 & 0 \end{bmatrix} \times \begin{bmatrix} 1 & 0 & 1 \\ 0 & 1 & 0 \\ 1 & 0 & 1 \end{bmatrix}$$

5×5 – Image Matrix 3×3 – Filter Matrix

The convolution of 5*5 image matrix multiplies with 3*3 filter matrix is called "**Features Map**" and show as an output.

ISSN: 2249-7196
IJMRR/July 2023/ Volume 13/Issue 3/01-16
DR. K. SUDHAKAR/ International Journal of Management Research & Review

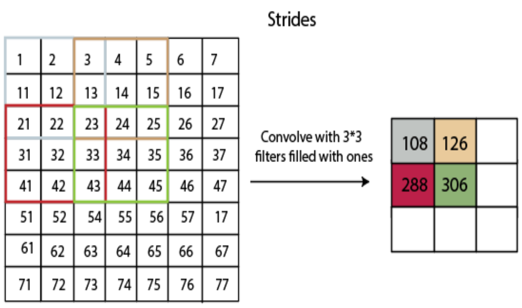
$$\begin{bmatrix}1&1&1&0&0\\0&1&1&1&0\\0&0&1&1&1\\0&0&1&1&0\\0&1&1&0&0\end{bmatrix} \times \begin{bmatrix}1&0&1\\0&1&0\\1&0&1\end{bmatrix} = \begin{bmatrix}4&3&4\\2&4&3\\2&3&4\end{bmatrix}$$

Convolved Feature

Convolution of an image with different filters can perform an operation such as blur, sharpen, and edge detection by applying filters.

Strides

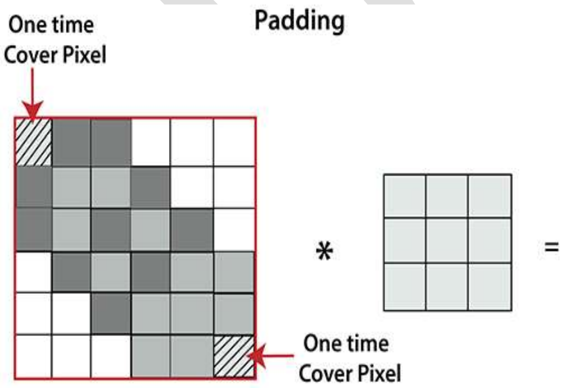
Stride is the number of pixels which are shift over the input matrix. When the stride is equaled to 1, then we move the filters to 1 pixel at a time and similarly, if the stride is equaled to 2, then we move the filters to 2 pixels at a time. The following figure shows that the convolution would work with a stride of 2.



Padding

Padding plays a crucial role in building the convolutional neural network. If the image will get shrink and if we will take a neural network with 100's of layers on it, it will give us a small image after filtered in the end.

If we take a three by three filter on top of a grayscale image and do the convolving then what will happen?



It is clear from the above picture that the pixel in the corner will only get covers one time, but the middle pixel will get covered more than once. It means that we have more information on that middle pixel, so there are two downsides:

- Shrinking outputs
- Losing information on the corner of the image.

To overcome this, we have introduced padding to an image. **"Padding is an additional layer which can add to the border of an image."**

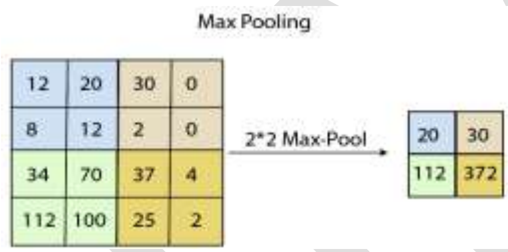
Pooling Layer

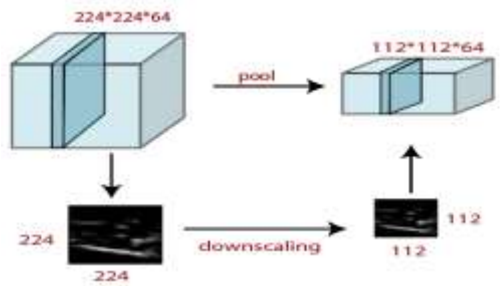
Pooling layer plays an important role in pre-processing of an image. Pooling layer reduces the number of parameters when the images are too large. Pooling is "**downscaling**" of the image obtained from the previous layers. It can be compared to shrinking an image to reduce its pixel density. Spatial pooling is also called downsampling or subsampling, which reduces the dimensionality of each map but retains the important information. There are the following types of spatial pooling:

Max Pooling

Max pooling is a **sample-based discretization process**. Its main objective is to downscale an input representation, reducing its dimensionality and allowing for the assumption to be made about features contained in the sub-region binned.

Max pooling is done by applying a max filter to non-overlapping sub-regions of the initial representation.





Average Pooling

Down-scaling will perform through average pooling by dividing the input into rectangular pooling regions and computing the average values of each region.

Syntax

layer = averagePooling2dLayer(poolSize)

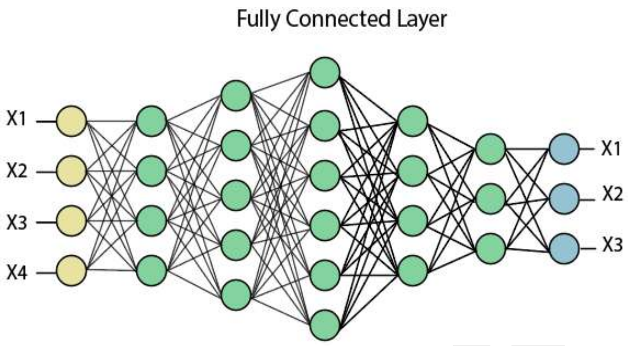
layer = averagePooling2dLayer(poolSize,Name,Value)

Sum Pooling

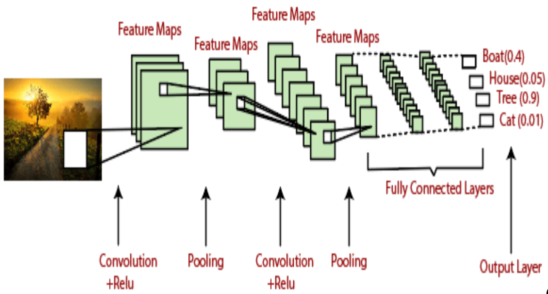
The sub-region for **sum pooling** or **mean pooling** are set exactly the same as for **max-pooling** but instead of using the max function we use sum or mean.

Fully Connected Layer

The fully connected layer is a layer in which the input from the other layers will be flattened into a vector and sent. It will transform the output into the desired number of classes by the network.



In the above diagram, the feature map matrix will be converted into the vector such as **x1, x2, x3... xn** with the help of fully connected layers. We will combine features to create a model and apply the activation function such as **softmax** or **sigmoid** to classify the outputs as a car, dog, truck, etc.



CONCLUSION In this work, we proposed an approach for estimating 2-D inter-frame displacements in the context of ultrafast US imaging. The approach consists of a CNN trained to restore high-quality images from single unfocused acquisitions and a speckle tracking algorithm to estimate inter-frame displacements from two consecutive frames only. Compared with conventional multi-acquisition strategies, this approach is immune to motion artifacts and enables accurate motion estimation at maximum frame rates, even in highly heterogeneous tissues prone to strong diffraction artifacts. Numerical and in vivo results demonstrated that the proposed approach is capable of estimating displacement vector fields from single-PW acquisitions accurately, including in zones initially hidden by SL and GL artifacts. The proposed approach may thus unlock the full potential of ultrafast US, with direct applications to imaging modes that depend on accurate motion estimation at maximum frame rates, such as shear-wave elastography or ultrasensitive echocardiography

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