

Emerging Wearable Ultrasound Technology

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Abstract—This perspective article provides a brief overview on materials, fabrications, beamforming, and applications for wearable ultrasound devices, a rapidly growing field with versatile implications. Recent developments in miniaturization and soft electronics have significantly advanced wearable ultrasound devices. Such devices offer distinctive advantages over traditional ultrasound probes, including prolonged usability and operator independence, and has demonstrated their effectiveness in continuous monitoring, non-invasive therapies, and advanced human-machine interfaces. Wearable ultrasound devices can be classified into three main categories: rigid, flexible, and stretchable, each having unique properties and fabrication strategies. Key unique strategies in device design, packaging, and beamforming for each type of wearable ultrasound devices are reviewed. Furthermore, we highlight the latest applications enabled by wearable ultrasound technology, encompassing continuous health monitoring, therapy, and human-machine interfaces. This article concludes by discussing the outstanding challenges within the field and outlines potential pathways for future advancements.

Index Terms—Wearable ultrasound, ultrasonic transducer, piezoelectric material

I. INTRODUCTION

FOR decades, ultrasound has been a cornerstone of clinical procedures due to its cost-effectiveness, non-invasiveness, safety, versatility, and convenience [1],[2]. Traditional ultrasound probes are bulky, operator-dependent, and confined to clinical settings for short durations [3]. Recent advancements in materials design and advanced fabrication have led to wearable ultrasound devices, offering low form factors, operator independence, and continuous monitoring even outside the hospital [4]. The low form factors are achieved with performance comparable to conventional ultrasound probes. Wearable devices can maintain a fixed position on the body without an operator to hold them, and thus eliminate operator dependency. Additionally, they can be worn for continuous and long-term monitoring without being limited by operator fatigue or requiring the subject to remain stationary for long durations. This rapidly emerging field has shown promise to be effective for continuous health monitoring [4],[5] delivering drugs [6], assisting wound healing [7], and facilitating human interfaces [8]. Wearable ultrasound technology can potentially transform healthcare and bring value-added benefits to various stakeholders at large, including patients, healthcare providers, and clinical researchers. This perspective article provides an overview of the current status of wearable ultrasound

technology, followed by sharing of our thoughts on several future directions.

II. MATERIALS AND FABRICATIONS FOR WEARABLE ULTRASOUND PROBES

A critical component of wearable ultrasound technology is the probe, which can be based on a single transducer or an array of transducers. There are three primary forms of wearable ultrasound probes: rigid, flexible, and stretchable. All three forms of probes share similar choices of transducers, but differ in their fabrication techniques, electrodes, and substrates, which give rise to their distinct mechanical properties.

A. Transducer Materials

Commonly used transducer materials in wearable ultrasound probes include piezoelectric ceramics, piezoelectric polymers, and micromachined ultrasound transducers. Piezoelectric ceramics typically refer to lead zirconate titanate and its composites (e.g., 1-3 composite), because of their high cost-efficiency. Other piezoelectric ceramics, such as lead magnesium niobate-lead titanate with a high electromechanical coupling coefficient [9] and lead-free lithium niobite, are

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TABLE I

PROPERTIES OF REPRESENTATIVE MATERIALS FOR WEARABLE ULTRASOUND TRANSDUCERS					
Material	Type	d_{33} (pC/N)	k_t^2	Z (MRayl)	Reference(s)
Bulk lead zirconate titanate	Piezoelectric ceramics	374-593	0.17-0.30	34.2	[10],[11],[12]
Lead zirconate titanate thin film	Piezoelectric ceramics	21	0.12	18.3	[10],[12]
1-3 Lead zirconate titanate/epoxy composite	Piezoelectric ceramics	>500	0.44-0.56	9.0-17.0	[13],[14]
Lead magnesium niobate-lead titanate	Piezoelectric ceramics	1143-2900	0.38	23.9	[15],[16]
Lithium niobite	Piezoelectric ceramics	7.9	0.32-0.41	29.3	[17],[18]
Polyvinylidene fluoride	Piezoelectric polymer	18-62	0.02	3.9	[19],[20]
Poly(vinylidene fluoride-trifluoroethylene)	Piezoelectric polymer	30-108	0.09	4.3	[20],[21],[22]
Poly-l-lactic acid	Piezoelectric polymer	5-12	<0.01	2.9	[23],[24],[25],[26],[27]
Capacitive micromachined ultrasound transducer	Micromachined ultrasound transducer	-	0.16-0.45	~1.5	[28],[29],[30]
Piezoelectric micromachined ultrasound transducer	Micromachined ultrasound transducer	-	0.15-0.25	~0.2-9.0	[31],[32]

seldom reported for wearable ultrasound applications due to their low cost-efficiency [33],[34]. Piezoelectric polymers typically refer to polyvinylidene fluoride and its copolymers, such as poly(vinylidene fluoride-trifluoroethylene), due to their high piezoelectricity. Biodegradable poly-l-lactic acid has also been reported as an implantable ultrasound transducer [35], but its low piezoelectricity limits general ultrasound applications. Micromachined ultrasound transducers include capacitive and piezoelectric types [36],[37]. The former offers a broad bandwidth yet requires a high bias voltage. The latter, despite its narrow bandwidth, is power-efficient, making it promising for wearable applications.

The transmitting efficiency and acoustic impedance are two of the most important properties for biomedical applications of ultrasound transducers. Transmitting efficiency represents the amount of acoustic energy generated by the transducer under a defined excitation level. Acoustic impedance determines the effective acoustic energy transmitted into the human body. Piezoelectric ceramics are known for having high transmitting efficiency and thus transmitting strong ultrasound waves, albeit encountering a considerable acoustic impedance mismatch with the human body. In contrast, piezoelectric polymers exhibit a reduced impedance mismatch, albeit at the expense of transmission energy. Micromachined ultrasound transducers

balance between these two characteristics. Table I gives a comprehensive comparison across piezoelectric ceramics, piezoelectric polymers, and micromachined ultrasound transducers.

B. Backing and Matching Materials

Backing and matching layers are integral parts of ultrasound transducers and can substantially enhance the transducer performance.

Without a backing material or when air-backed, the transducer delivers energy to both forward and backward media. The backward transmitted energy can be reflected at the transducer-air interface into the forward medium, increasing the total forward transmission energy and thus signal to noise ratio [38]. However, such reflection can happen multiple times, which elongates the spatial pulse length and reduces the transducer bandwidth and spatial resolution for pulse-echo applications. Backing layers can reduce the back reflection and the spatial pulse length. Existing wearable ultrasound probes, especially those designed for pulse-echo applications, often have backing layers that are made of metal-epoxy resin composites [4]. These materials have high conductivity, similar acoustic impedance to transducers, and high acoustic attenuation. Thus, they can function as electrodes, accept back-propagating energy, and effectively dissipates acoustic energy with the viscous resin

matrix [39]. In essence, backing layers trade the transmission energy for shorter pulse duration and thus better bandwidth and spatial resolution [40].

Matching layers can reduce the interfacial reflection. At each interface between the transducers and the skin, unwanted reflections and energy losses can occur due to these acoustic impedance mismatches. Strategic placement of matching layers between the transducers and the forward medium serves as a critical approach to minimize such interfacial energy loss. The matching layers are typically made of composites [40] or metamaterials [41]. Composites are often designed to have a thickness equivalent to a quarter of the wavelength, aligning the phases of the incoming and internally reflected waves within the matching layer. This phase synchronization enhances the transmission energy, subsequently boosting ultrasound signal propagation. The metamaterial-based designs introduce a gradient transition of acoustic impedance along the direction of ultrasound propagation, effectively eliminating interfacial energy loss. As a result, the matching layers enhance the ultrasound transmission power, and thereby increase the signal-to-noise ratio and penetration depth.

C. Transducer Fabrication

The primary concern for transducer fabrication is the transducer frequency, which is determined according to the trade-off between ultrasound propagation depth and spatial

resolutions. Low frequencies can propagate deeply, but have relatively low spatial resolutions, making them suitable for measuring deep targets such as the visceral organs. On the other hand, high frequencies are typically used for measuring superficial targets, such as the skin and gums.

The transducer frequency defines an upper bound for the required transducer inter-element pitch and the kerf (i.e., the gap between adjacent transducer elements). The pitch is crucial for determining the signal-to-noise ratio of an ultrasound probe. At a given active transducer area, where a certain amount of acoustic energy is transmitted, a smaller pitch can offer a better signal-to-noise ratio because more energy is converged to the main acoustic beam. Typically, a pitch smaller than half wavelength is adopted [42]. The kerf isolates an ultrasound transducer from the acoustic crosstalk brought by its neighboring elements. The larger the kerf, the better the acoustic isolation. However, a large kerf sacrifices the active transducer area and therefore the acoustic energy output, a more important transducer property. Thus, the general kerf design practice is to use the minimum manufacturable kerf [43].

For low-frequency applications (<15 MHz), piezoelectric ceramics are normally employed, benefiting from efficient mechanical dicing. Arrays of piezoelectric polymers [44],[45] or micromachined ultrasound transducers [46],[47] can also be developed, but those processes are typically intricate and expensive. For high-frequency applications (>15 MHz), pitches and kerfs become smaller. It is challenging to fabricate

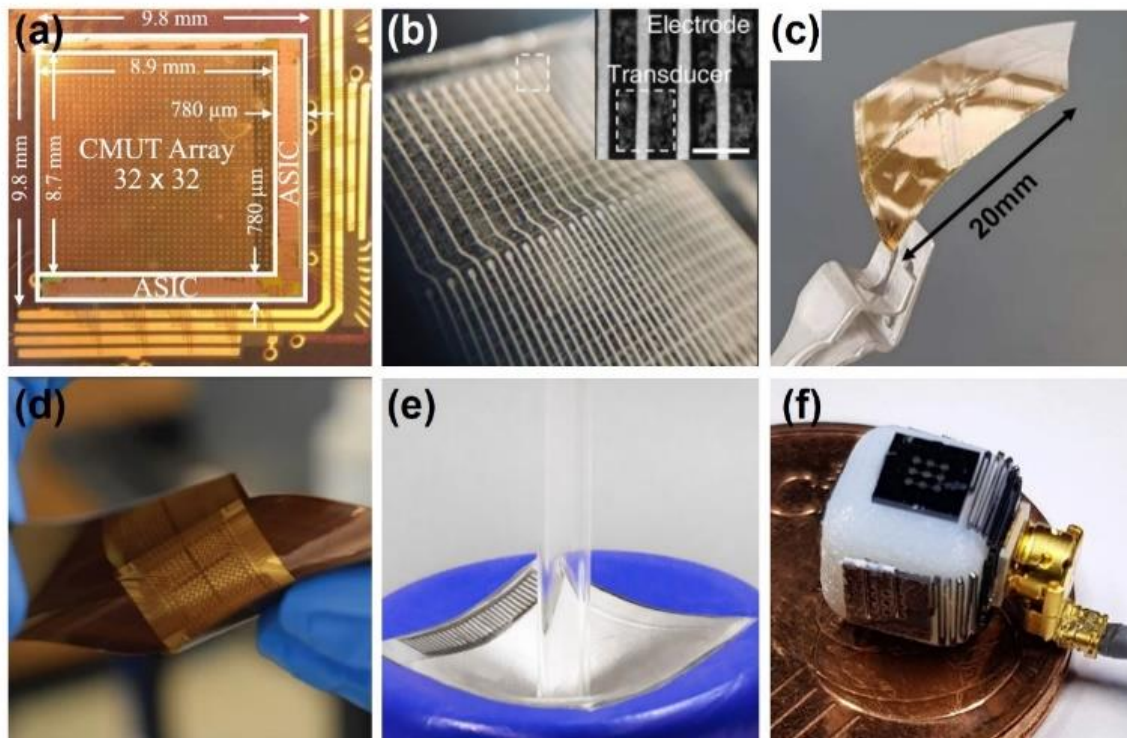


Fig. 1. The rigid, flexible, and stretchable wearable ultrasound probes. The rigid wearable ultrasound probes powered by capacitive micromachined ultrasound transducers on a printed circuit board (a) and lead zirconate titanate composite on an epoxy substrate (b). Adapted from [13] and [14]. The flexible wearable ultrasound probes enabled by piezoelectric micromachined ultrasound transducers on polyimide (c) and lead zirconate titanate on a flexible printed circuit board (d). Adapted from [15] and [16]. The stretchable wearable ultrasound probes enabled by lead zirconate titanate composite on styrene-ethylene-butylene-styrene (e) and piezoelectric micromachined ultrasound transducers on silicon (f). Adapted from [4] and [17].

piezoelectric ceramic arrays, due to limited resolution with mechanical dicing [48],[49]. For these applications, complementary metal-oxide-semiconductor processes employed in fabricating micromachined ultrasound transducers can offer very fine pitch and kerf.

Notably, the fabrication resolution is not associated with the creation of single transducer-based wearable ultrasound probes, where pitch is not of concern. In this case, piezoelectric ceramics and polymers have been extensively reported due to their straightforward and cost-effective fabrication methods [50],[51],[52],[53]. Micromachined ultrasound transducers remain underrepresented in single-transducer wearable ultrasound probes due to their relatively high cost.

D. Substrates and Electrodes

While different types of wearable ultrasound probes share common materials for the transducer, backing, and matching, a significant discrepancy exists in the choice of substrate and electrode materials that lead to different mechanical properties of the probes.

Rigid probes highlight the use of rigid substrates, such as epoxy [54], glass fiber-epoxy weave [55], and silicon [56], which are selected based on the electrode fabrication process being used. The electrodes are fabricated with high yield by well-developed methods, such as 3D printing and deposition-lithography [Fig. 1(a)] [55],[57],[58]. 3D printing features an inexpensive and easy operation. However, it has limitations in speed and may encounter filament breaks for line widths below $\sim 100 \mu\text{m}$, limited by the nozzle size. Deposition-lithography employs metal electroplating/sputtering, mask development, and etching to construct electrode patterns, offering scalability, reliability, and a much finer line width, but at a higher cost, than 3D printing. To interface with external control electronics, the piezoelectric ceramics and polymer transducers need to be bonded using wires or curable conductors [Fig. 1(b)] [54]. For micromachined ultrasound transducers, the electrodes and control electronics can be fabricated together with the transducers because they share the same complementary metal-oxide-semiconductor processes [56]. Rigid probes are often bulky and constraining user activities. Particularly, bulkiness becomes a problem when the probe aperture is large, which leads to a huge air gap between the probe and the human body. The air interfaces can cause pronounced reflections, and thus ultrasound gel or hydrogel [54] is necessary to fill these gaps. The gel evaporates gradually and needs frequent replenishment. Despite the bulkiness, rigid probes are particularly suitable for high-frequency applications because the involved fabrication processes are compatible with small pitches and kerfs.

Flexible probes are less bulky than rigid probes and thus offer an enhanced user experience. Flexible probes are built on polymer substrates, such as polyvinyl chloride, polyethylene terephthalate, and polyimide [50],[51],[59],[60],[61],[62],[63],[64]. Polyvinyl chloride is toxic and has a low melting point ($\sim 60^\circ\text{C}$), making it unsuitable for wearable applications due to transducer-generated heat during prolonged use. Polyethylene terephthalate is non-toxic

and has a higher melting point ($\sim 260^\circ\text{C}$) than polyvinyl chloride. Notably, polyimide has the highest melting point ($\sim 380^\circ\text{C}$), and is also non-toxic and more flexible than polyethylene terephthalate [Fig. 1(c)] [59]. Notably, hydrogel can also be used as a flexible substrate [54]. Hydrogel is non-toxic and provides better flexibility due to its higher pore concentration and water content than other common flexible substrates [65]. However, hydrogels cannot be used as a substrate for fabricating ultrasound transducers, because they can only tolerate a low temperature ($<40^\circ\text{C}$) [66],[67],[68]. For electrodes on flexible substrates, such as silver ink and alloys [56],[62], the fabrication is more complex than that on rigid substrates. Plenty of fabrication technologies are available for fabricating silver ink electrodes [51]. Despite the limited fabrication options, alloys have a conductivity that is one order of magnitude higher than that of silver ink. Alloys take advantage of mixed metal components to reach a balanced performance. For example, in titanium/silver alloy, titanium provides high tensile strength and corrosion resistance while silver provides high conductivity. These electrode thin films can sustain bending due to their small thickness and thus negligible strain. The fabrications of flexible probes are generally underdeveloped with lower yields than rigid probes, relegating them primarily to low-frequency applications [Fig. 1(d)] [69],[70].

However, mere flexibility is suboptimal for human body integration because the probe cannot conform to the skin during skin stretching. Stretchable probes solve this problem, resulting in ultrasound gel-free measurements and negligible constraints to user activities. This stretchability is achieved by using an “island-bridge” structure encapsulated in an elastomer. When selecting such an elastomer, its stretchability, bonding strength with transducers and electrodes, and moisture permeability for protecting electronic components are critical. Common choices include silicone elastomers such as polydimethylsiloxane [71],[72],[73] and EcoFlex [74],[75],[76],[77], and thermoplastic elastomers such as styrene-ethylene-butylene-styrene [4]. Polydimethylsiloxane excels in bonding strength ($\sim 0.3 \text{ MPa}$) [78], but is less stretchable ($\sim 100\%$) [79] and relatively moisture permeable (water vapor transmission rate $\sim 400 \text{ g/m}^2/\text{day}$) [80]. EcoFlex possesses a high stretchability ($\sim 600\%$) [79] and low moisture permeability (water vapor transmission rate $\sim 125 \text{ g/m}^2/\text{day}$) [81], but poor bonding strength ($\sim 20 \text{ kPa}$) [82]. Styrene-ethylene-butylene-styrene has a superior stretchability ($>2000\%$) [83] and bonding strength ($\sim 1 \text{ MPa}$) [84], but high moisture permeability (water vapor transmission rate $\sim 1900 \text{ g/m}^2/\text{day}$) [85]. Different use cases will have to prioritize different aspects of these elastomers.

Stretchable electrodes are more challenging to fabricate than those on rigid and flexible substrates. They are typically manufactured based on structurally-enabled stretchable conductors, filler-based stretchable conductors, or liquid metals [Fig. 1(e)]. The structurally-enabled stretchable conductors are relatively low-cost but unsuitable for dense electrodes in high-frequency ultrasound devices due to their space-consuming structures [Fig. 1(f)] [71],[74],[75],[86]. The other two types of stretchable electrodes are more costly but space-efficient,

suitable for dense electrodes. Filler-based stretchable conductors are doped elastomers, such as silver nanowires doped polydimethylsiloxane [72],[73]. Electrodes based on liquid metals, such as eutectic gallium-indium, can be fabricated by screen-printing or laser ablation [4]. Despite the high cost, liquid metals have more fabrication options and a higher stretchability than doped elastomers. Filler-based stretchable conductors and liquid metals may introduce an additional level of stretchability by structural design [87],[88]. Like flexible probes, stretchable probes face fabrication challenges and exhibit pitch variations during stretching, limiting their high-frequency applications.

Note that the impedance of wearable ultrasound probes tends to be higher than conventional rigid probes, because the wearable probes have smaller footprints and thinner interconnects, causing an increase in the wire resistance and transducer impedance. Thus, the matching circuits play an important role in matching the impedance of the probe with the transmission circuit for maximum transmission efficiency and signal-to-noise ratio of the system. In practice, the load impedance of the probe is often matched to offset its reactance (i.e., the imaginary part cancelled by the complex conjugate), so that the electrical energy can be converted to mechanical vibration rather than consumed by the probe reactance [89]. If the impedance is not properly matched, the reactance can restore and return delayed energy to the transmitter circuit and the transducer, causing ringing in the transmission line and reverberation of the transducers [40],[90], and thus lowering the signal-to-noise ratio.

III. BEAMFORMING STRATEGIES

As wearable ultrasound shifts towards increasingly complex multi-element devices, flexible and stretchable ultrasound devices encounter significant beamforming errors due to unknown array element positionings. Because the precision to which the element positions must be known is directly proportional to the wavelength, this is especially the case for high-frequency devices [91],[92]. Traditional rigid devices maintain fixed array element positionings, commonly consisting of flat and evenly spaced elements, and thus have a known delay profile for standard beamforming approaches such as delay-and-sum. In contrast, flexible and stretchable devices can conform to various dynamic surface geometries, which leads to unknown array element positionings, and consequently inaccurate delay profiles that cause beamforming aberration. Several studies utilizing various strategies for correcting such aberrations have been conducted. These strategies can be classified into two major types: making the surface geometry known through measurement, and finding the best solution to the unknown delay profile through algorithmic means.

In the first type of approach, a major strategy involves acquiring the 3D morphology of the surface to which the device is applied. Obtaining the 3D morphology can be achieved through various techniques such as 3D scanning [29] or MRI imaging [30]. However, the required external hardware

increases the operator dependency and can be costly. Moreover, variations in surface morphology due to factors such as changes in posture, breathing, satiation, and body habitus can cause the delay profile obtained using this approach to lose its accuracy over time, necessitating periodic updates to remain accurate. Alternatively, another strategy entails integrating a shape-sensing mechanism within the device itself. This mechanism can detect device deformation and identify array element positions in real time to reduce beamforming errors caused by changes in the surface without the need for external scanners. For instance, a shape-sensing optical fiber is capable of sensing strain distribution along the fiber via Rayleigh scattering from inherent defects [31]. However, it adds bulkiness to the device and is limited to measuring deformations along a one-dimensional sensor, rather than the whole device surface [93],[94].

In the second type of approach, optimization algorithms or artificial intelligence-based solutions can deduce the optimal delay profile using only radiofrequency data collected from an unknown surface [95],[96]. This strategy carries a higher computational burden but does not need any external hardware. Since the surface is always treated as an unknown variable in this approach, it is unaffected by changes in surface morphology, enhancing adaptability and reducing user dependence. An illustrative example of this approach treated the unknown surface geometry as an inverse problem. Image sharpness was evaluated to compare the beamforming results of various simulated time delay profiles, ultimately determining the optimal profile [32]. In another study, deep learning networks were trained on pairs of radiofrequency data and their corresponding correctly beamformed images, resulting in an end-to-end neural network capable of automatically determining the best delay profile based on a given set of radiofrequency data [33]. However, a limitation of using deep learning networks is the initial requirement for substantial data to train the network. Furthermore, for complex surfaces that necessitate more than two orthogonal functions for representation [97], this type of approach can fail to correct the beamforming aberrations, because the algorithm's objective function may become trapped by local maxima instead of converging on the global maximum [97],[98].

IV. WEARABLE ULTRASOUND FOR CONTINUOUS HEALTH MONITORING

Wearable ultrasound adopts multiple modes from traditional ultrasound for health monitoring, including A-mode, M-mode, B-mode, Doppler, and elastography. A-mode measures interfaces in media by recording the time-of-flight of ultrasound waves. M-mode concatenates echo signals in A-mode and visualizes interfacial changes over time. B-mode differentiates qualitative structural information of tissues by mechanical properties, such as density and stiffness. Doppler ultrasound estimates the speed of tissue movements and blood flow based on the frequency shift in echo signals. Elastography quantitatively measures tissue stiffness. All modes of traditional ultrasound probes experience operator dependence and limited usage. An experienced sonographer is necessary to correctly place and handle the traditional probe during scanning.

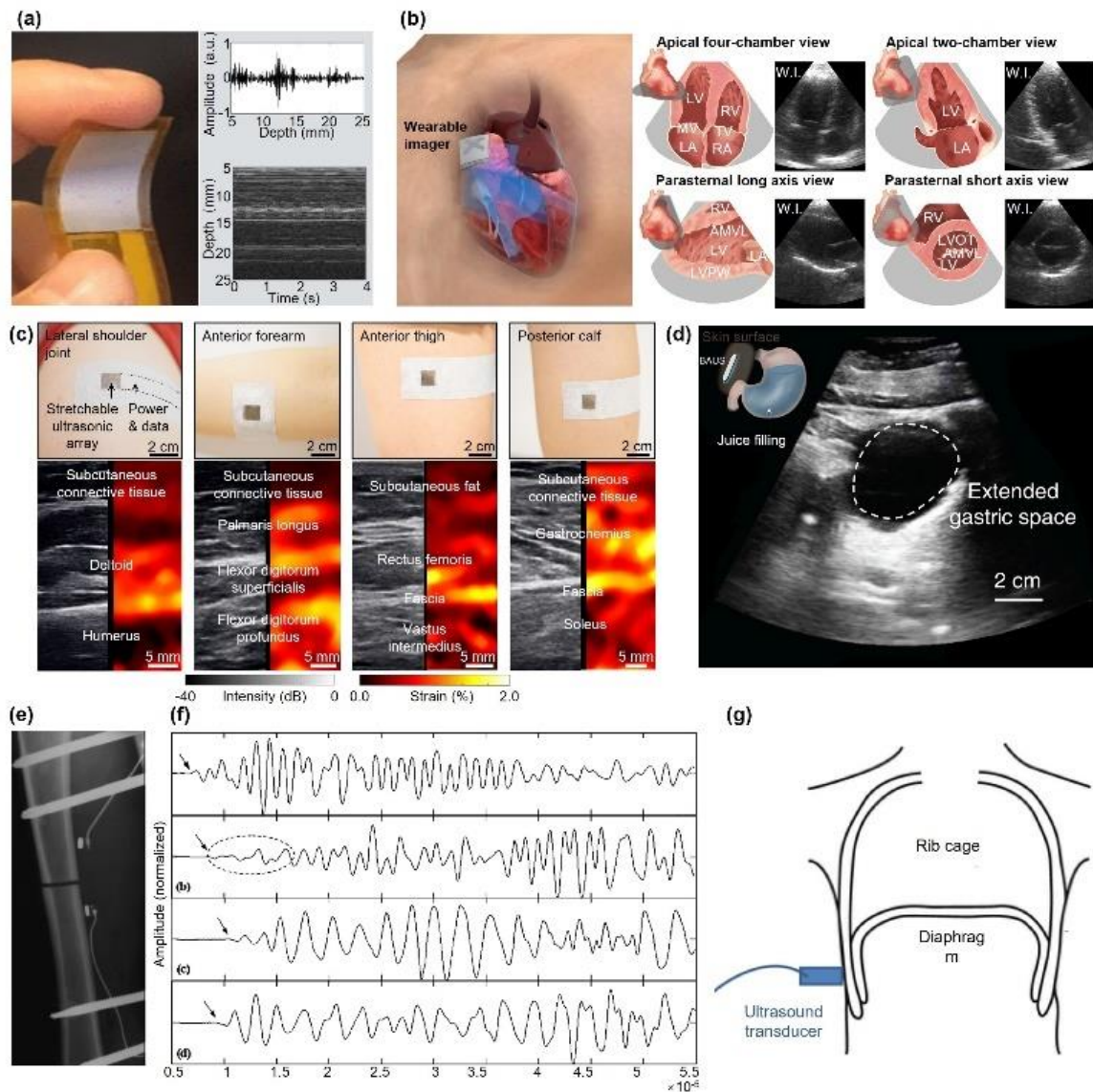


Fig. 2. Wearable ultrasound-enabled monitoring applications. (a) A wearable ultrasound device measures the common carotid artery using A-mode and M-mode. Adapted from [64]. (b) Cardiac images from four standard views taken by a soft and stretchable wearable cardiac B-mode imager. Adapted from [4]. (c) A stretchable ultrasound device monitors the muscle modulus of, from the left panel to the right panel, shoulder, forearm, thigh and calf. The corresponding B-mode images and elastographies are attached below each optical photo. Adapted from [39]. (d) A wearable ultrasound device monitors gastric juice filling. Adapted from [14]. (e) A wearable ultrasound transmitter-receiver pair attached to fractured bones and (f) its A-mode signals. From top to bottom are the A-mode signal collected before, right after, six weeks after, and thirteen weeks after fracture. The arrows indicate the first-arriving ultrasound pulse used for propagation velocity determination. The dotted circle indicates a low-amplitude wave pattern that does not present in later healing stages. Adapted from [73]. (g) A schematic depicting a diaphragm monitoring setup where an ultrasound transducer adheres to the skin over the rib cage. The diaphragm and the skin maintain stable positions relative to the rib cage. Adapted from [32].

The need for an operator leads to limited ultrasound scans with short durations, and unavailability outside clinics. The low form factor and hands-free operation of wearable ultrasound probes overcome the two limitations, leading to operator-independent continuous health monitoring of various physiological systems.

A. Cardiovascular Monitoring

The cardiovascular system requires continuous monitoring due to its dynamic activities. Underlying information such as blood pressure, blood flow, contractility, plaque presence, vessel stiffness, and cardiac output can be inferred from monitoring

the pulsatile behavior. Because the pulsatile behavior and blood flow could be monitored using A-mode, M-mode, and Doppler ultrasound in single-transducer setups [5],[76],[77],[99],[100],[101],[102],[103],[104],[105],[106],[107], the barrier to entry of wearable ultrasound-based cardiovascular monitoring devices was relatively low. Recent work enabled by advances in fabrication has produced wearable transducer arrays, enabling more powerful B-mode imaging [4],[54].

Major blood vessels, such as the carotid artery or jugular vein, offer easy targets due to their large diameter and accessibility,

and can provide a wealth of information about the central vascular system through single-transducer A-mode, M-mode [Fig. 2(a)], and Doppler ultrasound setups [5],[76],[77],[99],[100],[101],[102],[103],[104],[105],[106],[107]. Similar information can also be obtained from monitoring the peripheral vascular system via the radial and ulnar arteries in the wrist [34],[108],[109], which is more convenient for users, but may not be as diagnostically valuable as central arteries. For example, the peripheral blood pressure may not be as indicative of cardiovascular health as the central blood pressure [110]. Such devices can also monitor the cerebral blood flow using direct transcranial approaches [111], or indirectly from the carotid artery [99],[106]. Recent innovations have targeted the heart [4],[51],[54],[112], which poses unique challenges due to its great tissue depth, required imaging window size, fast motion, complex structures, and obstructions from bone and lungs. Early attempts resulted in wearable devices that could only track myocardial depth using A-mode [5],[112]. The collected A-mode signals only contained the time-of-flight information along a single direction from a single transducer.

Significant breakthroughs emerged with wearable Doppler and B-mode imaging of the heart [4],[54],[77]. B-mode imaging represented a significant advancement beyond single transducer A-mode and M-mode. The use of arrays necessitated a large number of ultrasound elements to be fabricated, and dense networks of electrodes to connect them. Furthermore, with multiple elements involved, beamforming algorithms were needed to reconstruct the images, introducing higher demands in both hardware and software. Notably, Fig. 2(b) shows B-mode images of four standard cardiac views taken by a soft and stretchable wearable ultrasound patch [4].

B. Musculoskeletal Monitoring

Wearable ultrasound technology can offer real-time and accurate assessments of musculoskeletal health, send warnings before injuries happen, and provide long-term monitoring of recovery processes without restricting body motion.

Contractile properties of muscles are associated with fatigue and atrophy [113],[114],[115],[116]. Monitoring the relative change in muscle thickness is a reliable estimator of contractile properties, because muscles tend to maintain their volume during motion and thus muscle shortening has an inverse relationship to its thickness [117]. The stronger the muscles contract, the thicker they become. Extensive research has demonstrated the potential of wearable ultrasound technology in monitoring muscle contractile properties using A-mode on muscle-mimic phantoms [55],[60],[61],[118],[119]. In-vivo tests on the gastrocnemius muscle [60],[119],[120], triceps muscle [61], and forearm [121] have further validated the feasibility of monitoring during motion in A-mode. Similar applications were demonstrated on the index finger and forearm using M-mode [50]. However, A-mode and M-mode are not accurate because a single echo signal can only estimate the muscle distribution. Two-dimensional B-mode images could give a comparatively more accurate estimation on

muscle-mimic phantoms [55],[118]. Muscle modulus has been shown to be associated with muscle contractile property [122]. Recently, stretchable ultrasound elastography allowed three-dimensional accurate mapping of muscle modulus and thus muscle contractile properties [Fig. 2(c)] [74]. The mapping results have been validated against magnetic resonant elastography.

The bones do not embrittle as easily as muscles fatigue and are less prone to injury. As a result, bone monitoring is usually only required post-injury during the recovery phase. Wearable ultrasound devices, equipped with A-mode and configured with a transmitter-receiver pair attached to fractured bone segments [Fig. 2(e)], could continuously monitor bone healing during fracture management [123]. The devices leverage transosseous ultrasound propagation to monitor bone healing stages by analyzing the mean propagation velocity and signal amplitude. Right after a bone fracture, mean propagation velocity decreases due to a slower sound speed in soft fillings between bone segments, such as collagen and inflammatory cells. Meanwhile, a lower signal amplitude is observed because of the acoustic impedance mismatch between the bone segments and the soft fillings. As healing progresses, a higher mean propagation velocity and signal amplitude are present due to osteogenesis [Fig. 2(f)]. While traditional ultrasound requires several visits, continuous bone healing monitoring provides real-time feedback to patients and reduces their dependence on echo labs [124]. Furthermore, wearable ultrasound technology could facilitate medical interventions in bones. For example, wearable B-mode images could help identify spinal structures, crucial for guiding needle operation during lumbar puncture [125], thereby freeing the interventionists' hands for more challenging tasks.

C. Other Monitoring Applications

The versatility of wearable ultrasound monitoring is evident across various domains, including fetal development, urinary sensation, respiratory pattern, and numerous others.

Pregnancy is a complex process with potential complications for both the mother and the fetus. During the nine-month gestational period, both the mother and fetus are in a delicate state where a single complication at any time can lead to adverse outcomes if left unchecked. Abnormal fetal movements, heart rates, and uterine contractions are often associated with poor birth outcomes [126],[127],[128], which make adequate monitoring of the mother and fetus crucial, especially in low-/middle-income countries with limited access to healthcare and prenatal services [75]. Cardiotocography is an established method for monitoring pregnancies, which combines Doppler ultrasound and tocodynamometers to monitor the fetal heart rate and uterine contractions. However, cardiotocography requires trained operators and bulky equipment that further exacerbates the inconvenience of pregnancy. Furthermore, traditional cardiotocography can frequently encounter false alarms due to motion of the mother or loss of the fetal heart rate from probe motion. Wearable ultrasound devices reduce the

operator dependency, can maintain a stable position on the body, and can be used for continuous monitoring. They have the potential to significantly impact pregnancy outcomes, particularly in developing countries where routine checkups may not be available. Most solutions for wearable monitoring have relied on strapping the devices to the body using cumbersome straps [126],[129],[130], which constrain the mother's abdomen and may not be suitable in the long term. The latest efforts have focused on miniaturizing these devices and developing self-adhesive, flexible ultrasound devices for simultaneous monitoring of maternal and fetal signals [128],[131]. These devices employ traditional rigid ultrasound transducers housed in flexible silicone shells with movable joints, making them more portable and user-friendly. Other sensors such as photoplethysmogram and tocodynamometer can be housed together for comprehensive monitoring of both maternal and fetal parameters.

The urinary system filters waste by-products from the blood, producing urine and storing it in the bladder before excretion. Unconscious urinary incontinence, a prevalent issue affecting children, the elderly, and individuals with spinal cord injuries, can significantly impact the affected individuals' quality of life. The unawareness of full-bladder sensation can lead to unintended urine leakage, posing social concerns. Real-time estimation of bladder volume is a key application where wearable ultrasound technology can make a significant impact. A-mode ultrasound has been demonstrated to provide real-time volume estimation of the bladder on phantoms [57],[132], leading to successful commercialization [57],[125],[133]. However, the accuracy of A-mode ultrasound signals was compromised by deviations in the ultrasound beam's incidence angle during breathing, potentially causing inconsistent measurements. The time-of-flight measured in A-mode varied along with the deviations, resulting in a false-increasing or decreasing bladder volume estimation. 3D B-mode ultrasound, though undeveloped, promises accurate volume reconstruction and addresses the uncertain incidence angle of A-mode.

Impairments in the respiratory system can lead to oxygen deprivation and irreversible damage. Given the unpredictable nature of respiratory disorders like sleep apnea, continuous monitoring of the respiratory system is crucial. Wearable ultrasound devices combining A-mode and B-mode have effectively detected obstructive sleep apnea by monitoring tongue base deformation, which can indicate pharyngeal airway closure [134]. More generally, these devices can measure the respiratory rate by observing the diaphragm and pleura positions using M-mode and B-mode [54],[62],[135]. Compared to photoplethysmography that is widely adopted for respiration monitoring, wearable ultrasound technology has shown superior reliability. Photoplethysmography indirectly measures respiratory activities by reading oxygen saturation based on light absorption [136]. It can produce erroneous readings due to muscle-induced arterial movements affecting light absorption, i.e., motion artifacts. Wearable ultrasound technology, by directly pinpointing the diaphragm, overcomes these artifacts [62]. Unlike the case of bladder monitoring, the diaphragm's position relative to the ultrasound transducer

remains consistent given that the rib cage directly connects to the diaphragm and secures the skin via muscles and connective tissues [Fig. 2(g)]. With the rib cage acting as a constraint, breathing does not result in ultrasound beam deviation; instead, it induces only parallel sliding with respect to the diaphragm's surface. This movement has minimal impact on its localization, effectively avoiding motion artifacts. This aspect is crucial for the anticipation of respiratory failure, a leading cause for admissions to intensive care units [137].

Wearable ultrasound technology thrives in many other health-monitoring areas. For instance, a flexible ultrasound device can serve as an implantable cranial sensor, providing valuable data for long-term post-operative monitoring of brain tumor regrowth [138]. A similar device can also continuously monitor eye-blinking patterns [139], offering insights into neurological conditions. In cancer screening, wearable ultrasound devices have been developed for early screening and diagnosis of breast cancers [33],[140]. Integrating wearable ultrasound devices with hydrogel adhesives has enabled long-term imaging of various abdominal organs in B-mode, such as the stomach [Fig. 2(d)] [54]. These examples exemplify how wearable ultrasound technology extends its scope in health monitoring.

V. WEARABLE ULTRASOUND FOR THERAPIES

Wearable ultrasound technology harbors the potential as a therapeutic tool, by leveraging the ultrasound-induced effects such as sonophoresis, cavitation, and heating. The non-invasive, unobtrusive design of wearable ultrasound devices ensures a comfortable user experience while maintaining reliable performance in therapeutic interventions.

A. Ultrasound-triggered Drug Delivery

Ultrasound has emerged as a promising noninvasive alternative to traditional needle-based transdermal drug delivery administration [Fig. 3(a) and 3(b)]. This innovative technique harnesses sonophoresis, wherein ultrasound waves modulate the permeability of the skin surface. The key mechanisms involved are the acoustic cavitation and streaming effects. The cavitation causes microscale gas bubbles to form, grow, oscillate, and collapse. Notably, the shock waves generated during the collapse of these bubbles play a crucial role in enhancing skin permeability. By microscopically disrupting the stratum corneum, these shock waves facilitate the efficient transport of drugs and other substances through the skin barrier. The streaming effect introduces continuous radiation forces to the tissue, which can lead to localized particle displacements and fluidic currents. The radiation force can move particles in the direction of the propagating ultrasound waves, and thus drug molecules can be delivered into the tissue [141].

Wearable ultrasound devices could effectively enhance the transdermal delivery of various chemicals, including insulin [6], niacinamide [142], and hyaluronic acid [143]. These advancements hold immense promise in improving patient comfort and adherence to medication regimens, especially for individuals with needle phobia and those requiring frequent

injections. Continued research and development in this field has the potential to offer a safe, efficient, and patient-friendly approach, revolutionizing conventional drug delivery.

B. Ultrasound-assisted Wound Healing

Ultrasound treatment has demonstrated its effectiveness in promoting wound healing, particularly in chronic wounds. Fibroblast senescence plays a critical role in chronic wound recovery [7]. Notably, ultrasound stimulation has been found to activate Rac1 protein in fibroblasts, reversing fibroblast aging and effectively promoting chronic wound healing [7]. Preclinical studies have shown that ultrasound accelerates wound healing in both dermal and epidermal layers [7]. Additionally, low-intensity ultrasound (0.3 to 1 W/cm²) has been observed to enhance wound healing through cavitation and streaming effects. These effects contribute to increases in cell membrane permeability, facilitating the diffusion of cellular metabolites. Additionally, a relatively higher intensity ultrasound (1~1.5 W/cm²) allows for cavitation and thermal effects. These not only enhance membrane permeability and molecular diffusion but also raise local tissue temperature, promoting in-tissue remodeling and healing [144],[145],[146].

Traditionally, ultrasound wound dressing has been limited to optimizing frequency and power to promote healing. Wearable devices allow long-term and continuous effects, introducing time as a new knob for optimizing treatment outcomes. Yet, a comprehensive understanding and safety regulations for

ultrasound-assisted healing must be developed accordingly. Current studies focus on correlating ultrasound activation and changes in metabolism rate in general [147]. There are no studies to elucidate the mechanism behind such changes. This leaves significant gaps in our understanding of how ultrasound energy impacts cellular activities, gene expression, or molecular behaviors. For safety, therapeutic ultrasound currently borrows regulations from diagnostic ultrasound [148]. However, because diagnostic ultrasound aims to minimize tissue interaction, whereas therapeutic ultrasound seeks to maximize it, the appropriateness of such shared regulations is highly questionable. This ambiguity may hinder further development in therapeutic ultrasound, necessitating distinct safety guidelines designed to meet its unique needs.

VI. WEARABLE ULTRASOUND FOR HUMAN-MACHINE INTERFACES

Ultrasound waves demonstrate exceptional penetration in tissues, rendering them an ideal modality for interfacing deep tissue signals and machinery. Utilizing wearable ultrasound devices, we can continuously record skeletal muscle activities for robotic or prosthetic control, and record or modulate neural activities, thereby presenting unprecedented opportunities for creating effective human-machine interfaces.



Fig. 3. Wearable ultrasound enabled drug delivery and human-machine interfaces. (a) A conformable ultrasound patch for cavitation-enhanced transdermal drug delivery (left) and the design of a conformal ultrasound patch (right). Adapted from [102]. (b) A stretchable ultrasound facial mask for sonophoresis. Adapted from [103]. (c) A wearable transcranial ultrasound stimulation device on a freely behaving mouse. Adapted from [104]. (d) A flexible ultrasound patch for peripheral nerve stimulation. Adapted from [105]. Implementing ultrasound-based neural interface in a mouse (e) and a rabbit (f). Adapted from [8] and [106].

A. Ultrasound for Controlling Robots and Prosthesis

Wearable ultrasound has emerged for controlling robots and prosthetic devices in an unobtrusive and long-term manner. Through A-mode sensing, these devices can accurately track morphological changes in deep muscles and tendons. The changes in tissue interfaces induced by different gestures can adjust the propagation path of the ultrasound waves. Therefore, the pulse-echo signals are encoded with information about the tissue interfacial distributions specific to each gesture [149],[150]. Prior studies have validated the viability of assessing changes in tissue interfacial distribution using wearable ultrasound devices on the forearm and calf, enabling gesture estimation [149] and quantification of muscle contractions [60]. In contrast to surface electromyography, ultrasound offers a more direct and accurate measurement of deep muscle morphology [149]. This offers a more comprehensive and reliable grasp of user motion, thereby enhancing control precision.

Additionally, air-coupled low-frequency ultrasound devices have proven effective in non-contact tracking of lip movements or gestures. Researchers have demonstrated that integrating ultrasound transmitters and receivers with glasses enables the recognition of speech at natural speaking speeds and styles [96]. The transmitters generate ultrasound waves, and the echoes reflected from the lips are collected by the receivers. The echo patterns, encoded with lip motions, can be analyzed by a machine learning model to infer speech. In a similar fashion, an array of ultrasound transducers mounted on the wrist can detect motions of fingers and palm from echoes transmitted through air [151]. Through machine learning, these gestures can be classified and recognized with a high accuracy rate of 92.5%.

B. Ultrasound-mediated Neural Interfaces

Efficient neural interfaces pave ways toward better understanding of the neural circuits, which holds implications for numerous biomedical applications, including motor rehabilitation [152], neurological disorder treatment [153], and memory/cognitive enhancement [154]. However, achieving efficient neural interfaces poses two main challenges for existing wearable devices (e.g., electromagnetic or optics devices). First, tissue attenuation dampens the power substantially, which limits detectable depth. Second, signal fidelity is compromised owing to tissue scattering during signal propagation. Wearable ultrasound technology offers a unique solution to overcoming these challenges. Ultrasound waves provide a highly efficient and reliable link for delivering power and transmitting data between the wearable receivers and the neural circuits. For power delivery, the wearable device can generate ultrasound waves to resonate the implanted ultrasound transducers and thus convert the mechanical energy into electrical power [8]. For signal transmission, ultrasound waves can achieve signal communication between the wearable devices and the implants through the amplitude modulation method [8].

Recent studies have demonstrated that wearable ultrasound devices could modulate brain circuits in animals [Fig. 3(c)] and stimulate nerves in humans [Fig. 3(d)]. For animal studies,

mice are of particular interest as their thin skulls have minimal ultrasound attenuation. The wearable stimulation probes were miniaturized and wirelessly controlled to maintain the animal's mobility. In freely moving mice, the brain was continuously stimulated while the cerebral hemodynamic responses were monitored [155]. Therefore, a stimulation-monitoring closed-loop was formed to better understand the function of the brain circuits. In humans, a flexible ultrasound device was designed to stimulate the peripheral nerves, such as the vagus and tibial nerves [156], which could help treat conditions like epilepsy and various psychiatric disorders.

In addition, wearable ultrasound has been used for power delivery and data communication with implanted modules. For power delivery, the wearable transducers would generate focused ultrasound beams and the energy could be collected by the transducers in the implanted module, which converts the mechanical energy to electrical energy [157]. Compared to other deep tissue power delivery methods, such as electromagnetics and optics, ultrasound has low tissue attenuation and is independent of the implanted coil size, thereby offering high-efficiency [158],[159]. For communication, wearable transducers can fine tune receiving directivity through beamforming, allowing backscattered signals to be received with high fidelity. For example, neural signals can be encoded as the impedance load by the control electronics of the implanted module. Consequently, changes in neural signals alter the module's impedance load, affecting the amplitude of backscattered echoes received by the wearable transducers [160]. Preclinical studies on rodents [8] and rabbits [161] [Fig. 3(e) and 3(f)] have demonstrated the effectiveness of ultrasound in these applications.

Moreover, ultrasound-mediated visual prostheses have exhibited remarkable progress in restoring vision [162]. A transducer array is directly connected to stimulation electrodes, which activate the retina when the transducer produces electrical responses to external acoustic pressure. Prosthetics utilizing this mechanism have demonstrated substantial potential in restoring visual sensations, thereby holding immense promise for enhancing the quality of life for individuals with visual impairments [162].

VII. CIRCUIT INTEGRATION

Despite these advancements in wearable ultrasound devices, many still require connecting the front-end transducers to bulky backend driving hardware, restricting practical portable use. Standalone devices with on-board processing and wireless communication can drastically improve device portability, expand access to ultrasound technologies, and promote internet of medical things.

Recent advances in integrating wireless chips have made it possible to create both pulse-echo [Fig. 4(a)] and continuous wave Doppler systems [Fig. 4 (b)]. For the pulse-echo system, a multiplexer, a transmit/receive switch, a pulse generator, a receiver, a sequencer, an analog-to-digital converter, and a wireless (Wi-Fi) module could be integrated to support arrayed

ultrasound probes [Fig. 4(c)]. This system could reconstruct the anatomical structure of the tissue and monitor dynamic changes at the tissue interfaces. Multiple imaging modes including A-mode, B-mode, and M-mode could be achieved [5]. For the continuous wave Doppler system, a clock generator, a transmitter amplifier, a low-noise receiver, a quadrature demodulator, filter circuits, and audio modules are integrated [Fig. 4(d)]. Although this is a non-imaging system, the

hemodynamic parameters, such as blood flow velocity and volume, could be accurately measured [101].

VIII. AUTOMATED DATA PROCESSING BY ARTIFICIAL INTELLIGENCE

While continuous data streams can be collected with the use of wearable devices, they still need to be read by a trained expert. Manually interpreting the large amounts of continuously generated data places a heavy burden on the interpreter. Thus,

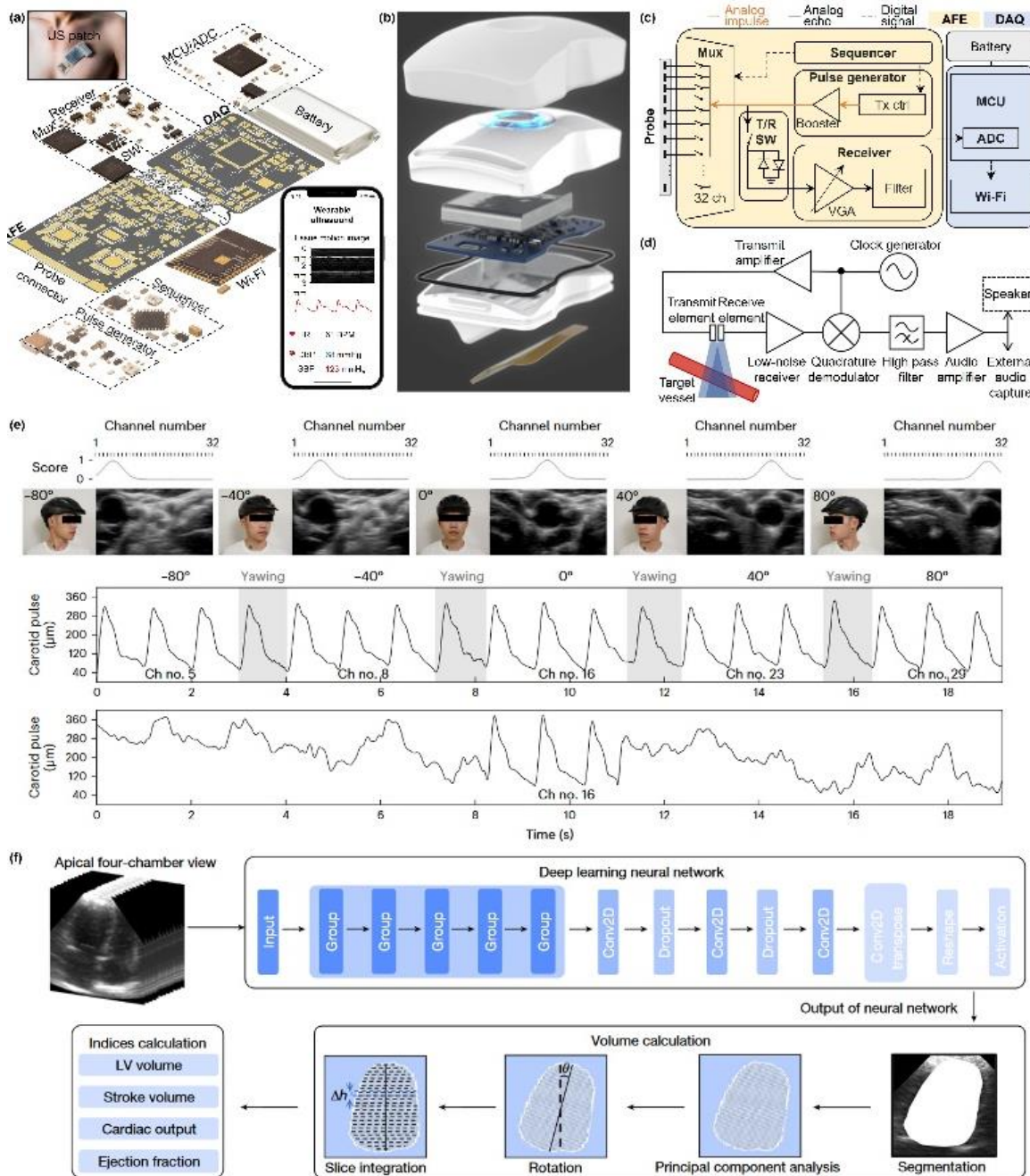


Fig. 4. Circuit and algorithm integration for wearable ultrasound devices. (a) Exploded view of the pulse-echo system. Inset: the ultrasonic patch on chest for cardiac activity monitoring. Adapted from [5]. (b) Exploded view of the continuous wave Doppler system. Adapted from [62]. (c) Schematic diagram of the wireless pulse-echo ultrasound circuit. Adapted from [5]. (d) Schematic diagram of the continuous wave Doppler circuit. Adapted from [62]. (e) The prediction score of the transducer channel for tracking the carotid artery (top) is used to continuously record the carotid artery pulse waveform during head yawing (middle), while the signal is lost if only using one channel (bottom). Adapted from [5]. (f) Cardiac images are segmented by a deep-learning neural network, and the geometry of the segmented image can be analyzed to determine quantitative cardiac indices. Adapted from [4].

although wearable devices allow user-independency of the data collection, the data interpretation still remains a major obstacle standing in the way of user-independence.

The big data made available by wearable ultrasound creates unprecedented opportunities for the development of machine learning algorithms, which could automatically interpret the data, freeing up the work of physicians and making the devices viable for use by the general population. Existing work has already begun to demonstrate the utility of automated data processing. For example, a pair of wearable sensors placed on the abdomen monitored the lung volume and respiratory pattern. Machine learning was used to predict the posture of the subject, including standing and lying on the back, left, and right [58]. Another investigation used a wearable ultrasound device to track the pulsation of the carotid artery. During rolling, pitching, and yawing of the head, the device could continuously collect the pulsation waveform without interruption, by using machine-learning to select the optimal moving sub-aperture containing the signal [Fig. 4(e)] [5]. In another study, a deep learning model was used to segment cardiac images from a wearable ultrasound device. The segmented images could then be analyzed to derive continuous waveforms of cardiac indices such as stroke volume, ejection fraction, and cardiac output [Fig. 4(f)] [4].

IX. DISCUSSIONS AND PERSPECTIVES

Wearable ultrasound technology has experienced rapid growth, fueled by advancements in materials design, fabrication techniques, and data processing algorithms. It offers advantages such as extended usage and operator independence, driving novel applications. Wearable ultrasound devices have been proven to be effective in continuous monitoring, non-invasive treatment methods, and advanced human-machine interfaces. However, several challenges persist.

There have been numerous fabrication techniques for wearable ultrasound probes, but the challenge of achieving a long lifetime persists, especially for flexible and stretchable probes. These probes' substrates expose solder sites to stress during deformations such as bending and stretching, accumulating fatigue in soldering materials. The fatigue results in cracks and therefore an open circuit after repeated deformations, diminishing the probe's lifetime. Advances in fabrication techniques, such as localized structural reinforcement of solder sites [163],[164], holds potential for extending probe longevity.

Owing to the high frequency and megahertz-level bandwidth of the ultrasound probes, the wearable ultrasound circuitry requires a high sampling rate and processing speed. Thus, the power consumption is intrinsically high. Additionally, the substantial data output from wearable ultrasound is hampered by limited data throughput rates, especially in wireless devices. Future innovations should prioritize the power efficiency of the entire system to extend battery lifetime. Integrating low-power application-specific integrated circuits and efficient power harvesters could potentially significantly enhance the battery lifetime of fully integrated wearable ultrasound systems [5],[101]. Implementing local processor or application-specific

integrated circuits could provide local computational power to mitigate the high data throughput. The onboard computing units could perform local beamforming to reduce the data load for wireless transmission.

Artificial intelligence can analyze the long-term patterns from continuous data streams enabled by wearables to uncover patterns that humans cannot, based on complexity and sheer size of the data [165]. Further studies in high-efficiency data labeling or unsupervised learning algorithms are needed to efficiently process these large datasets into usable training data for artificial intelligence models [166]. In this way, artificial intelligence training could be greatly streamlined, guiding the evolution toward automated data processing, results interpretation, and decision-making. In the future, it may be possible to completely eliminate artifacts, noise, shadows, and image distortions, uncover new patterns hidden in the continuous data streams, or even perform preliminary diagnoses without the assistance of a physician. Advancements in artificial intelligence have the potential to elevate wearable ultrasound technology to a new level of independence and monitoring accuracy, reducing operator reliance, and improving precision.

The widespread adoption of wearable ultrasound in clinical practice holds the promise of discovering novel applications uniquely enabled by this technology, such as continuously monitoring and performing real-time tests during exercise [4],[5]. However, there is limited validation of its clinical value over existing approaches. Therefore, clinical trials are still needed to understand the added benefits of continuous monitoring or real-time testing during exercise. In addition, thorough comparisons of the advantages offered by wearable ultrasound devices over the gold-standard approaches are necessary to assess the trade-offs between accuracy and reliability of the gold-standard versus convenience, long-term usage, and non-invasiveness of wearable ultrasound in real-world situations. Such comparisons in monitoring blood pressure have already been conducted [5], and should become the standard practice as wearable ultrasound technology matures and outspreads.

In its early stages, wearable ultrasound has been typically limited to probing the mechanical properties of tissues. With the miniaturization of ultrasound into increasingly compact forms, alongside the miniaturization of other sensor types, the integration of ultrasound with multiple sensing modalities in single wearable devices becomes highly feasible. Some studies have already integrated wearable ultrasound with temperature sensing [126], cardiocography [131], photoacoustic sensing [167],[168], or chemical sensing [104], transcending the limits of the individual modalities. As the field continues to grow, we anticipate exciting discoveries and transformative innovations that will continue to improve healthcare outcomes and enhance the quality of life for individuals worldwide.

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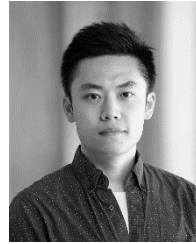
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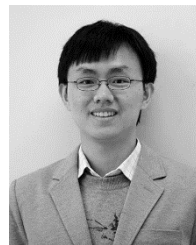
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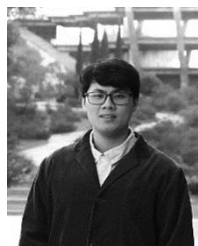
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