Flexible Ultrasonic Transducers for Wearable Biomedical Applications: A Review on Advanced Materials, Structural Designs, and Future Prospects

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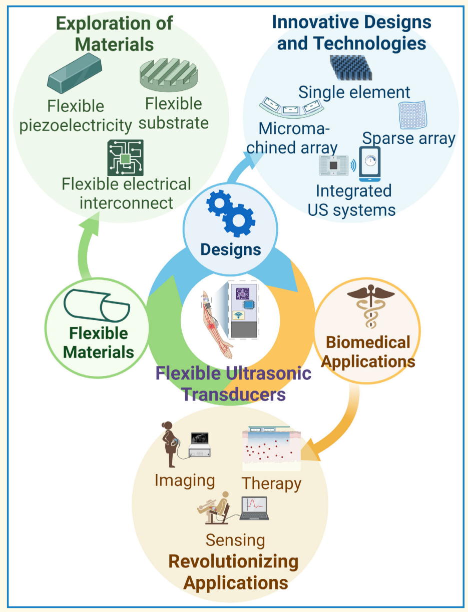
***Abstract*—Due to the rapid developments in materials science and fabrication techniques, wearable devices have recently received increased attention for biomedical applications, particularly in medical ultrasound (US) imaging, sensing, and therapy. US is ubiquitous in biomedical applications because of its noninvasive nature, nonionic radiating, high precision, and real-time capabilities. While conventional US transducers are rigid and bulky, flexible transducers can be conformed to curved body areas for continuous sensing without restricting tissue movement or transducer shifting. This article comprehensively reviews the application of flexible US transducers in the field of biomedical imaging, sensing, and therapy. First, we review the background of flexible US transducers. Following that, we discuss advanced materials and fabrication techniques for flexible US transducers and their enabling technology status. Finally, we highlight and summarize some promising preliminary data with recent applications of flexible US transducers in biomedical imaging, sensing, and therapy. We also provide technical barriers, challenges, and future perspectives for further research and development.**



***Index Terms*—Biomedical imaging, biomedical sensing, biomedical therapy, flexible electronics, flexible ultrasound, ultrasound transducers.**

I. Introduction

ULTRASOUND (US) imaging and sensing are frequently used diagnostic modalities in biomedical research and clinical practice [1], [2]. By virtue of acoustic impedance mismatch in various tissues, US waves can detect and distinguish biological structures, such as vessels, muscles, and bones. According to the morphological changes in different biological structures, various diseases and health statuses can be diagnosed and monitored [3]. For example, US can measure blood pressure fluctuations based on artery deformation [4], [5], evaluate muscle fatigue due to muscle thickness change [6], muscle contractility changes via US-derived strain measurements [7], [8], [9], and voluntary muscle torque via muscle geometrical features [10], [11], [12], and identify brain region activation in the light of blood volume differences [13], [14]. Traditional US transducers, whether employing single-element configurations for sensing

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Highlights

- **Unveils cutting-edge flexible ultrasound tech, spotlighting its role in imaging, sensing, and therapy, prioritizing materials, fabrication, and applications.**
- **Surveys materials, designs, fabrication, biomedical applications, and outlines research barriers, guiding future development in flexible ultrasound transducers.**
- **Providing a comprehensive overview for researchers in the biomedical community, propelling the field towards enhanced diagnostic capabilities and transformative healthcare solutions.**

or multielement arrays for imaging, typically possess rigidity and bulkiness that pose challenges for wearable biomedical applications on the skin. Furthermore, prevalent US imaging and sensing instruments, such as US probes, often necessitate the expertise of a sonographer to conduct ultrasonic diagnoses. Compounding these issues, the rigid and planar nature of traditional US probes introduces a geometric disparity when applied to the curved surface of human skin, leading to inadequate acoustic conditions at the testing interface. This inability to tightly conform to the skin further hinders their performance. The above limitations in terms of hardware and software inhibit the translation of healthy monitoring from "short term" to "long term" from "at-hospital" to "at-home."

Nowadays, recent advances in flexible devices, including flexible electrical circuits, flexible substrates, and flexible components, enable the development of wearable US biomedical applications, paving the way for future long-term disease diagnosis and health monitoring at home [15]. By using island-bridge electrodes encapsulated with silicone elastomer, scholars integrated multiple rigid elements into a stretchable transducer array to achieve satisfactory US imaging performance on intricate surfaces, enabling the successful and ongoing assessment of waveform patterns related to central blood pressure and deep-tissue hemodynamics [16], [17], [18]. Instead of using rigid elements, silver nanowire-based stretchable electrodes mixed with polydimethylsiloxane (PDMS) can also achieve excellent flexibility for one 1-3 composite single-element transducer. Scholars have utilized this kind of flexible single-element US transducer to sense blood pressure fluctuations [19], [20]. In addition, continuous monitoring of patient physiological signals using US transducers prefers the assistance of dry US couplant, not gel couplant. Recently, scholars developed a hydrogel-elastomer hybrid couplant that is soft, stretchable, and bioadhesive [21]. This innovative device is believed to boost the development of long-term continuous US imaging and sensing.

US transducers have been the subject of numerous studies and reviews over the years. The chronological order of these reviews starts with Brown's paper [22] in 1992, which reviewed piezoelectric polymer US transducer technology in medical and nondestructive testing applications. Zhou et al. [23] discussed the fundamental principles and design factors related to piezoelectric ultrasonic transducers, along with their applications in the field of biomedicine. Chan et al. [24] presented the development and challenges of capacitive micromachined US transducers (CMUTs) for Photoacoustic imaging. Finally, Yang and Zhang [25] conducted an overview of recent progress in advanced flexible devices utilizing inorganic thin films, encompassing technologies such as micromachined ultrasonic transducers (MUTs). Chen et al. [26] reviewed the design methods of piezoelectric US transducers (PUTs), presenting optimization design methods and discussing their future perspectives. He et al. [27] conducted a review of recent advancements in piezoelectric materials for PMUTs, summarizing their prototypes and applications during the last decade and discussing future development.

Several review papers have highlighted the potential of flexible US transducers in biomedical imaging applications. For instance, La et al. provided an overview of progress in high-performance materials, fabrication technologies, and medical imaging applications [15]. Similarly, Ren et al. [28] reviewed recent advancements in functional materials optimization, transducer designs, and medical imaging applications. Jiang et al. [29] explored the use of wearable US bioelectronics for continuous healthcare monitoring, emphasizing their potential in personalized health management. It discusses various categories of flexible bioelectronics, with a focus on wearable US imagers for cardiac imaging [29]. Lin et al. [30] examined the advancement of soft wearable devices tailored for deep-tissue sensing, facilitating the acquisition of health-relevant physiological signals. Their exploration encompasses diverse sensing methodologies, encompassing US transducers, device designs, and performance criteria [30]. Chen et al. [31] provided an extensive review of recent advances in flexible wearable sensors designed for continuous monitoring of cardiovascular vital signs. These sensors encompass a range of mechanisms, including ultrasonic technology, and are categorized into five distinct groups. The review comprehensively explores their materials, working principles, and potential medical applications, emphasizing the critical role of early detection and management in combating cardiovascular diseases [31]. However, a more comprehensive and forward-looking review is needed to cover recent advancements in both materials and structural designs, as well as their potential for a range of biomedical applications.

The proposed review article aims to offer a fresh perspective on the current state of the art in materials and structural designs for flexible US transducers. While other reviews have touched on this topic, this article will provide a more comprehensive overview, placing particular emphasis on the potential biomedical applications of these devices, including imaging, sensing, and therapy. This article will cover a broad range of topics, including the advanced materials and optimized fabrication methods used to produce flexible US transducers, as well as their integration and structure. It will also examine the challenges and future perspectives for research and development in this field. Our review stands out with its comprehensive coverage of flexible US transducers. We explore enriched content across sections: "Flexible Piezoelectric Materials" delves into various materials and includes a dedicated comparison table; "Structure of Flexible Transducers" expands to include single-element transducers, sparse arrays, flexible laser-generated US transmitters, and wearable and integrated ultrasound system. For each type of flexible ultrasound transducer, we have introduced new subsections dedicated to their Role in Biomedical Applications. We delve into the suitability of these transducers for various biomedical applications, presenting a nuanced understanding of their advantages and disadvantages. This approach provides readers with fresh insights into how to select the most suitable flexible ultrasound transducer when confronted with real-world biomedical scenarios; "Flexible Ultrasound Transducers in Biomedical Applications" encompasses a wide range of applications; "Challenges and Future Perspectives" extensively discusses topics such as optimal components and wearable integration. This distinctiveness underscores our paper's innovative contribution, providing a comprehensive outlook on flexible ultrasound transducer technologies. This unique perspective will be of particular interest to researchers and practitioners in the biomedical imaging and sensing community, as well as those working in ultrasound therapy. In addition, this article will provide a detailed discussion of recent advancements in materials and structural designs, offering valuable insights into the future of flexible ultrasound transducers. Overall, this review article will serve as a comprehensive guide for anyone looking to gain a deeper understanding of the current state of the art in flexible ultrasound transducers and their potential for biomedical applications.

II. Exploration of Materials for Flexible Ultrasound Transducers

In the realm of flexible transducers, prior research has summarized the materials and structures [15], [30]. However, it is noteworthy that a comprehensive review addressing the categorization of material applications based on the underlying structure has been lacking. Hence, within this section, we endeavor to provide a concise introduction to various materials, culminating in a comprehensive overview of the choices available for the material of flexible transducers and arrays. To guarantee flexibility when conforming to the shape of the body, the flexible US transducers usually include the vital components of: 1) advanced materials or structures for converting electrical energy into mechanical energy and the reverse; 2) a highly deformable substrate for rendering flexibility of the US transducer; and 3) flexible electrical interconnect for maintaining the conductivity of each component during with deformation [15]. Though not necessary, flexible electrical interconnection is usually required for maintaining the conductivity of each component during deformation. In Section II-A, we will commence by introducing the materials utilized in flexible US transducers. Moreover, we will explore different classifications of flexible US transducers. Some flexible piezoelectric materials, substrates, and electrical interconnects are shown in Fig. 1.

A. Flexible Piezoelectric Materials

Currently, flexible piezoelectric materials dominate in all the flexible US transducers. A piezoelectric material exhibits the piezoelectric effect, which converts mechanical motion into electrical charge and *vice versa* [32]. As a fundamental element in US transducers, the choice of piezoelectric materials primarily hinges on factors such as piezoelectric coefficient, dielectric constant, electromechanical coupling coefficient, mechanical quality factor, acoustic impedance, and material flexibility. Favorable characteristics for piezoelectric materials used in flexible US transducers include a high electromechanical coupling coefficient (k) and a low acoustic impedance (Z_a) [16]. High k values are desirable for generating a larger electrical charge for a given amount of mechanical stress, which results in a stronger US signal. A low Z_a of piezoelectric materials represents a high transmission efficiency due to a favorable impedance matching with skins. To realize the flexibility of the piezoelectric materials, there are mainly two methods: 1) using the stretchable piezoelectric materials as the active layer directly [33] or 2) dispersing a flexible polymers matrix into the rigid piezoelectric materials [34]. The comparison of electromechanical properties for flexible active materials is summarized in Table I.

1) Flexible Piezoelectric Polymers: Compared with traditional piezoelectric materials, piezoelectric polymers have the advantages of lower acoustic impedance (around 4 MRayl) for matching body issues, a broad bandwidth, and permanent flexibility. The most commonly used piezoelectric polymer is polyvinylidene fluoride (PVDF) [33]. The first literature report on the use of PVDF in soft tissue imaging was by Ohigashi et al. [40] in the 1980s, in which a 5-MHz single-element PVDF transducer was mounted on a spherically concave copper backing for imaging of thyroid tissue. Since then, the PVDF has been applied as the polymer transducer for breast imaging [47] and tissue characterization with a center frequency of around 15 MHz [48]. The other polymer material that has been commonly applied in US imaging was PVDF-trifluoroethylene (PVDF-TrFE) [40], [41], [49], which improved the temperature stability and electromechanical coupling coefficient of PVDF. However, the comparatively low electromechanical coupling coefficients of <0.3 of the polymer materials compromised performance in the diagnostic frequency range [33]. The effects of temperature, frequency, and film orientation should be taken into account when designing PVDF-based ultrasonic transducers [50].

In recent advancements within the field of flexible piezoelectric polymer materials, two distinct strategies have arisen to overcome the constraints of traditional choices for medical implant applications. One approach involves the development of biodegradable piezoelectric nanofibers using poly(L-lactic acid) (PLLA) as the base material. This innovation enables the creation of biocompatible and

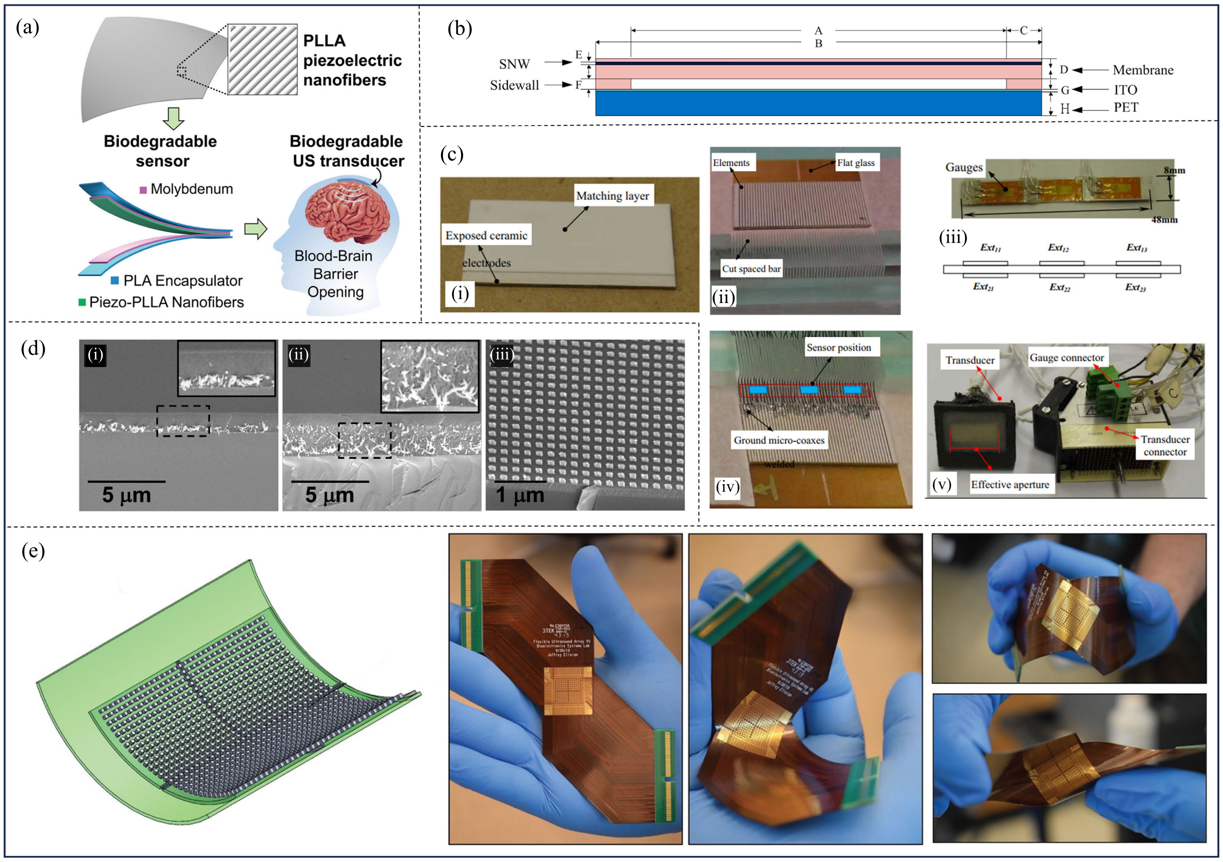

Fig. 1. Flexible piezoelectric materials, substrate, and electrical interconnect. (a) Demonstrating the potential of PLLA nanofibers with precise controllability and remarkable piezoelectric performance, offering a promising pathway for the advancement of biodegradable implanted piezoelectric devices. Adapted from [35]. (b) Cross-sectional portrayal of a transparent CMUT with an ITO-PET substrate of 125 μm thickness. Adapted from [36]. (c) Outline of the manufacturing process for the transducer utilizing PU rubber as the backing. Adapted from [37]. (d) Scanning electron microscopy (SEM) imagery of multi-walled carbon nanotube (MWCNT) films synthesized through chemical vapor deposition (CVD), including a PDMS overcoat, with varying CVD growth times: (i) 1-min growth time, (ii) 3-min growth time, and (iii) SEM depiction of an gold nanoparticles (AuNP) array before PDMS coating. Adapted from [38]. (e) 3-D representation of the flexible array featuring integrated piezoelectric transducers (left). Flexible printed circuit board (PCB) circuit with a diagonal connector design allowing for various bending modes: neutral position, easy hand bending, convex bending, and shear bending (right). Adapted from [39].

TABLE I
COMPARISON OF ELECTROMECHANICAL PROPERTIES FOR FLEXIBLE ACTIVE MATERIALS

Materials	Soundspeed (m/s)	Density (g/cm^3)	Z_a	K_t	Dielectric constant	Mechanical flexibility	Biocompatibility	Ref
PVDF	2200	1.78	3.9	0.15-0.2	6	Outstanding	No	[33], [40]
P(VDF-TrFE)	2400	1.88	4.5	0.3	5	Satisfactory	No	[41], [42]
PLLA	2280	1.25	-	0.064 (k_{14})	2.47	Satisfactory	Yes	[35], [43], [44]
Glycine-PCL	-	1.1145	-	-	-	Satisfactory	Yes	[35]
ZnO film on polyimide	-	-	-	0.24	9-12	Poor	No	[1], [45]
AlN film on Kapton	-	-	-	0.6	8-10	Poor	No	[46]
PZT-5H/PDMS 1-3 composite	3700	5.08	18.8	0.74	1007	Poor	No	[19]
PZT-5A/PDMS 1-3 composite	3480	4.41	15.1	0.7	433	Poor	No	[20]

efficient piezoelectric PLLA nanofibers that exhibit stable and controlled piezoelectric performance. Leveraging this breakthrough, novel applications such as biodegradable pressure sensors for monitoring physiological pressures and ultrasonic transducers for facilitating blood–brain barrier opening have been demonstrated, expanding the potential for drug delivery and medical device integration within the human body [see Fig. 1(a)] [35]. Another pioneering strategy focuses on the utilization of amino acid crystals, specifically glycine, as a piezoelectric material embedded within polycaprolactone (PCL) nanofibers. This approach results in the fabrication of flexible, biodegradable, and piezoelectric glycine-PCL nanofiber films that exhibit superior performance characteristics. The resulting piezoelectric glycine-PCL nanofiber film surpasses existing biodegradable transducers by demonstrating remarkable piezoelectric efficiency, notably with a high US output, thus opening new avenues for applications in medical implantation [51].

The comprehensive material options for flexible piezoelectric polymers used in transducers encompass a diverse range. Notably, PVDF, as reported in various studies [33], [40], [47], [48], stands out for its high flexibility. Enhancing the temperature stability and electromechanical coupling coefficient of PVDF, PVDF-TrFE has been explored as well

[40], [41], [49]. For biocompatible applications, PLLA has been studied, offering biodegradable piezoelectric nanofibers [35]. Equally impressive, PCL nanofibers have demonstrated remarkable piezoelectric efficiency among piezoelectric polymers, resulting in a notable increase in ultrasonic output [51]. It is important to note that, in comparison with conventional piezoelectric ceramics or crystals, most piezoelectric polymers exhibit lower piezoelectric coefficients, which may limit their utility as ultrasound transmitters [52].

2) Polymer-Coated Piezoelectric Ceramics: Compared with piezoelectric polymers, conventional bulk piezoelectric materials, including piezoelectric ceramics and crystals, are characterized by their high piezoelectric coefficients and high mechanical quality factors for US transducer applications. The primary challenge with using bulk piezoelectric materials in flexible applications is their inherent rigidity, which becomes even more significant in low-frequency ultrasound applications. One solution is polymer-coated piezoelectric ceramics, in which a piezoelectric ceramic is coated with a polymer layer to reduce the rigidity of conventional piezoelectric materials [53].

The advancement of flexible US transducers has involved the utilization of high-performance piezoelectric thin film materials, including aluminium nitride (AlN), zinc oxide (ZnO), and lead zirconate titanate (PZT) [1], [46], [53]. The primary challenge with using bulk piezoelectric materials in flexible applications is their inherent rigidity, which becomes even more significant in low-frequency ultrasound applications. These materials have the potential to be coated onto flexible substrates, such as thin metallic foils, leading to the creation of flexible US transducers. However, the PZT, AlN, or ZnO thin films may suffer from significant film stress and generate numerous defects during operation. Thus, the applications of these transducers are restricted due to their low durability. More than just controlling the thickness, researchers tried to address these challenges with porous films, in which the sol-gel spray technique was applied to produce PZT films on stainless steel and worked as the substrate and electrodes [1], [54]. The thin PZT/PZT films exhibit a notably lower electromechanical coupling coefficient compared to bulk PZT, which has implications for the efficiency of converting electrical and acoustic energy within the flexible active material. [1]. This phenomenon can be attributed to several key factors. First, the constrained nature of thin films, imposed by the substrate, limits their capacity to effectively convert electrical energy into mechanical vibrations and vice versa [55]. Moreover, interactions with the substrate can modify the domain structure of thin films, leading to a reduction in coupling efficiency. Surface effects, including charges and defects, can further disrupt the alignment of polarized domains, contributing to the observed lower coupling coefficient [56]. Finally, the inherent quality and imperfections of thin films can introduce challenges to the smooth operation of the piezoelectric effect, ultimately resulting in a diminished electromechanical coupling coefficient. Besides the limitation of the low k_t, the surface of the porous ceramic exhibited a high acoustic impedance mismatch with normal bio tissues, which limited its overall sensitivity as US transducer [45], [57], [58].

In a comprehensive overview of material choices for polymer-coated piezoelectric ceramics, it is evident that materials such as AlN, ZnO, and PZT have garnered significant attention. These materials possess the capability to be coated as thin films, granting them flexibility, which is a result of their inherent porosity and thinness. When paired with flexible coating substrates, such as thin metallic foils, the fabrication of flexible ultrasonic transducers becomes feasible. However, it is crucial to acknowledge that these materials exhibit a notably lower electromechanical coupling coefficient compared to their bulk piezoelectric counterparts. This reduction can be attributed to factors like surface defects, limited piezoelectric layer thickness, and inherent material quality imperfections. In addition, the materials face challenges related to acoustic impedance mismatch, which, in turn, impacts their overall sensitivity.

3) Piezoelectric-Polymer Matrix Composites: Flexible piezocomposites have garnered interest as a means to enhance the electromechanical coupling property and lower the acoustic impedance of flexible active materials. These composite structures comprise a combination of inflexible piezoelectric materials and pliable polymers. By combining these two materials, it becomes possible to adjust the composite's attributes, resulting in improved properties when contrasted with single-phase materials. The connectivity of different phases in the composite has been used to define the category of the piezoelectric composite [59], [60], [61], [62]. For the composite with two phases including polymer and piezoelectric materials, the 0-3, 2-2, and 1-3 composites are mostly used, in which the first and second numbers reflect the self-connected dimension of the piezo part and the polymer, respectively. Among these options, 1-3 piezocomposites, comprising rigid piezoelectric rods embedded within a polymer matrix, find frequent application in creating flexible US transducers. These composites possess thickness-mode electromechanical coupling coefficients that surpass those of rigid piezoelectric materials, approaching the values observed in the rod mode k_{33} of these materials. Furthermore, the substitution of a low-density, pliable polymer for dense and rigid piezoelectric materials can significantly decrease acoustic impedance. This arrangement effectively created a suitable distance between the compact rigid piezoceramic units within the flexible substrate, guaranteeing material flexibility. To design the 1-3 composite structures, both the theoretical model (effective medium model (EEM) [63]) and finite element analysis were applied with flexible 1-3 composite transducers reported for various US applications. For example, Harvey et al. [64] used piezoceramic (PZT-5A) fibers dispersed randomly in an epoxy polymer matrix to fabricate a 1-3 piezocomposite structure. Kim et al. [19], [65] introduced a flexible ultrasound transducer operating at 1.8 MHz, employing a PZT-5H/PDMS 1-3 piezocomposite as the flexible active material. The composite was prepared by a dice-and-fill method with PZT-5H and PDMS fillers [19], [65]. With a similar process, Peng et al. [20], [66] demonstrated a PZT-5A/PDMS 1-3 piezocomposite with a different volume fraction, resulting in an electromechanical coupling coefficient and the acoustic impedance of 0.68 and 13.37 MRayl, respectively.

In the realm of piezoelectric-polymer matrix composites, various piezoelectric materials, including PZT-5A, PZT-5H, and PMN-PT, have been combined with flexible polymers to create ultrasonic transducers. These composites offer advantages such as high flexibility, a broad bandwidth, and elevated sensitivity, which distinguish them from other flexible materials. Nonetheless, it is important to note that these composites may exhibit reduced flexibility compared to fully polymer-based materials. This is primarily attributed to factors like the height of the piezoelectric pillars and the size of the kerf, which are relatively larger in these composites. Furthermore, their fabrication process tends to be more intricate in comparison to purely polymer-based alternatives. This limitation is particularly pronounced in high-frequency composite applications where a minimal kerf is essential for effective performance [67].

B. Flexible Substrate

Flexible substates are usually used for the assembling of the flexible transducer or arrays and, thus, had the dominate impact over the flexibility of the whole structure [15]. Generally, this flexible substrate is crafted from deformable and flexible polymer materials or ultrathin metallic films, such as elastomers, soft polymer films, and stainless-steel films. These materials are chosen for their innate flexibility, elasticity, and ability to endure stretchability [68], [69]. The appropriate selection of flexible substrate material significantly influences the flexibility and endurance of US transducers.

Elastomers have emerged as essential materials for flexible and stretchable substrates in the realm of flexible ultrasound transducers. An example of a widely used silicone elastomer is PDMS [see Fig. 1(d)] [38], renowned for its high flexibility and impressive stretchability of over 170% in tensile strain. PDMS serves a dual role, serving as both a flexible substrate embedded with rigid piezoelectric material rods and as a protective packaging material for transducer devices.

Another notable substrate material is Ecoflex, a silicone elastomer commonly employed in flexible ultrasound applications. Differing from PDMS, Ecoflex exhibits a lower Young's modulus (Ecoflex-0030 (1:1) and PDMS (20:1) possess Young's moduli of approximately 60 kPa [70] and 1 MPa [71], respectively). This distinction led Hu et al. [16] to select Ecoflex as the substrate, superstrate, and filler in their device, ensuring a low modulus that facilitated seamless conformity to highly curved surfaces. In the subsequent research by Hu et al., they presented a fresh method that automatically aligns transducer elements with the bonding electrodes. The process is initiated by affixing a substantial backing layer to the 1-3 composite and then meticulously dicing the bonded bilayer into the intended array configuration. To prevent any tilting or shifting during bonding, a high-adhesion silicone elastomer, particularly Ecoflex-0030 (smooth-on), is employed to fill the kerf and unite individual elements. This careful and inventive manufacturing technique plays a critical role in guaranteeing the array's performance in challenging applications [72]. Polyurethane (PU) polymers were also used in flexible US transducers due to their high flexibility [37], [73]. In [37], the PU rubber was chosen as electrical isolation and mechanical protection for the active elements and backing layer for its great flexibility, resistance to fatigue, and also for having good acoustic damping [see Fig. 1(c)].

Another commonly used flexible substrate is a thermoplastic polymer, such as polyimide (PI), polyethylene terephthalate (PET) [36], [74], and polyethylene naphthalate (PEN) [75]. Introducing PET as a flexible substrate, Pang and Chang [36] demonstrated the creation of a flexible ultrasound transducer with an indium tin oxide-PET (ITO-PET) substrate [see Fig. 1(b)]. This transducer integrated SU-8 sidewalls, vibrating membranes, and a transparent electrode made of silver nanowires. Notably, this design exhibited remarkable visible-light transmittance exceeding 80% and the capability to operate on curved surfaces with a radius of curvature as tight as 40 mm. Similarly, in the context of flexible substrates, Kim et al. [75] presented an electrostatic actuator on a flexible PEN substrate. This innovative design emitted acoustic waves at ultrasonic frequencies, showcasing the adaptability and potential of PEN in flexible transducer applications. In comparison, PI emerges as another noteworthy flexible substrate due to its inherent characteristics [53]. PI, known for its beneficial attributes such as excellent biocompatibility, remarkable tensile strength, and inherent flexibility, has proven valuable in the construction of 2-D piezoelectric micromachined ultrasound transducer (PMUT) arrays when integrated with PDMS packaging. Such arrays, operating at a center frequency of 2.2 MHz, have been realized by researchers such as Yang et al. [76] and Wang et al. [58]. The exceptional mechanical properties of PI enable these arrays to perform effectively even under bending deformation, highlighting the advantages of this flexible substrate in enhancing the performance of ultrasound transducers. The properties of various substrates are compared in Table II.

C. Flexible Electrical Interconnect

The advent of flexible ultrasonic transducers, which merge flexible electronics and ultrasonic technology, is linked to the growth of miniaturized and flexible electronics. Conventional metal-based electrical interconnections had limitations in flexibility and stretchability, which are vulnerable to cracks or delamination under deformation [80]. To overcome these challenges, explorations over wearable or stretchable electronics have experienced a recent upswing [81].

Flexible electrical conductors are essential for the manufacturing of flexible US transducers, expanding their applications, as they preserve electrical conductivity even after undergoing mechanical deformation. In a single flexible US transducer, flexible electrical conductors such as silver nanowires serve as the electrodes of active materials [19], [20]. Flexible electrical conductors serve a dual purpose in flexible US arrays by functioning as both electrodes and interconnects. This facilitates the electrical connectivity of the active material elements or US transducer elements within flexible arrays [17], [82].

TABLE II
COMMON SUBSTRATES FOR FLEXIBLE US SENSORS

Material	Modulus [MPa]	Tensile strain [%]	Tensile strength [MPa]	Poisson's ratio	Ref.
Polyethylene terephthalate (PET)	2000 - 4100	60 - 165	22.0 - 95.0	0.3 - 0.45	[36], [74]
Polyimide (PI)	2500 - 10000	5 - 72	15.0 - 230	∼0.37	[77]–[79]
Poly(ethylene naphthalate) (PEN)	5000 - 5500	N/A	∼200	0.3 - 0.37	[75]
Polyurethane (PU)	10 - 50	100 - 396	55.2 - 62.1	0.48 - 0.49	[37], [73]
Polydimethylsiloxane (PDMS)	∼0.36 - 0.87	170 - 710	1.5 - 7.7	∼0.49	[19], [38], [65]
Ecoflex	<0.1	300∼700	∼1.3	∼0.5	[16], [17], [72]

There are two primary approaches to designing functional electrical conductor materials: using conductive polymers and using conductive composites made by incorporating a conductive phase into an elastomer matrix. Conductive polymers, such as poly(3,4-ethylenedioxythiophene) [83], poly(3,4-ethylenedioxythiophene) polystyrene sulfonate (PEDOT) [84], polyaniline (PANI) [85], [86], and polypyrrole (PPy) [87], have gained significant use as flexible electrodes in sensors and supercapacitors due to their notable flexibility and electrical conductivity. However, the utilization of conductive polymers as electrodes for flexible US transducers has been limited by their lower conductivity and electromechanical properties compared to other materials.

The use of conductive composites, composed of stretchable insulating elastomers and nonstretchable conducting fillers, is prevalent in stretchable electronics due to their high conductivity, high deformability, ease of printing, low cost, and simple processing methods [88]. These composites gain their stretchability from the mechanical stress–strain behavior of the insulating elastomer matrix, while their conductivity is determined by the percolation of the conducting fillers. According to percolation theory, there exists a threshold volume fraction of the fillers in the composite, at which the elastomer-filler composite undergoes a metal-insulating transition [89]. The composite is considered conductive when the volume fraction of the filler is equal to or greater than the percolation threshold, which is typically dependent on the form factor of the fillers, such as 0-D nanoparticles (e.g., carbon black [89] and Au nanoparticles [90]), 1-D nanowires [e.g., single-walled or multiwalled carbon nanotubes (CNTs)] [91], Ag nanowires [88], and 2-D nanosheets/nanoflakes (e.g., graphene and Ag flakes [92]). Most recently, liquid metal mixed with elastomers also has been used for flexible electrodes due to its higher flexibility (about 750% without failure) compared to traditional conductive composite [93].

An alternative method for creating flexible electrical conductors involves designing their geometric configurations using traditional rigid electrode materials. Electronic devices constructed on flexible plastic bases are often seen as having significant promise for use in unconventional electronics [94]. Geometric patterns such as ultrathin ribbons and serpentine-shaped filaments are commonly employed to craft flexible electrical conductors using stiff conductive materials. These structures can be manufactured through machining, printing, or microfabrication and nanofabrication. Metals, serving as exemplary rigid conductive materials, boast exceptional conductivity and find widespread utilization in the creation of these configurations for flexible electrical conductors. In order to achieve substantial mechanical deformations, a slender metal strip can be intricately fashioned into intricate patterns such as serpentine, meandering, or mesh-like structures. This design enables significant stretching capabilities while upholding the overall structural integrity. Employing a serpentine layout for the metallic traces strategically facilitates stretchability, achieved through a well-coordinated process of unfolding and bending at stress-release junctures. This innovative design strategy presents an avenue for achieving remarkable flexibility and deformability, suitable for diverse applications [94]. An illustrative example of the utilization of metals for flexible electrical interconnect can be found in [17]. In this study, a bilayer arrangement involving PI with a thickness of 4 μm and copper (Cu) measuring 20 μm was employed to create stretchable electrodes. This innovative approach was harnessed for the purpose of interconnecting a 4 $\times$ 5 array of transducers within the device. By employing this bilayer stacking technique, the researchers demonstrated an effective means of achieving both electrical conductivity and mechanical flexibility, thereby enabling seamless connectivity in a stretchable electronic system. Nevertheless, the effectiveness of flexible electrical conductors produced through this method is constrained by various elements, including the thickness of the ribbons or filaments, the adherence of these elements to the substrate, and the flexibility of the substrate itself. Furthermore, intricate configurations of flexible electrical conductors and the accompanying manufacturing procedures amplify the uncertainty and potential challenges associated with producing flexible US transducers.

III. INNOVATIVE DESIGNS AND TECHNOLOGIES FOR FLEXIBLE ULTRASOUND TRANSDUCERS

Flexible US transducers, akin to their rigid counterparts, manifest in various configurations, including single-element and array transducers. A foundational understanding of the structural components comprising typical US transducers is crucial in elucidating the innovation and technological advancements that enable their adaptability to diverse biomedical applications. This section delves into the structural intricacies of flexible US transducers, elucidating their pivotal role in the realm of biomedical ultrasound. Furthermore, it introduces various types of flexible US transducers and assesses their applicability to distinct clinical and research domains. Fig. 2 provides a visual representation of a standard

US transducer's anatomy, serving as a reference point for comprehending the ensuing discussions.

A. Flexible Single-Element Piezoelectric Transducer

1) Structure and Design of Flexible Single-Element Transducers: Conventional US imaging transducers are rigid because they are made of piezoceramics, piezocomposites, or piezoelectric single crystals. Compared to array transducers, single-element transducers have the advantage of being less complex, with a simpler fabrication process. One strategy for fabricating flexible single-element US transducers is using piezoelectric films, such as AlN, ZnO, PVDF, and PZT [45], [46], [77], [96], and flexible piezocomposites. Researchers have developed flexible single-element US transducers employing thin films of PZT or ZnO, featuring center frequencies of 13.1 and 29 MHz correspondingly [64]. Nonetheless, these transducers were manufactured without incorporating acoustic matching layers and backing blocks, factors that could potentially influence sensitivity and restrict bandwidth. Furthermore, the sensitivity or durability of the thin film transducers was also compromised [97]. As an alternative method, Taeyang et al. [19] introduced a flexible single-element US transducer using a flexible 1-3 composite. This approach involved employing a "dice and fill" technology to fabricate the device. The transducer featured a PZT-5H/PDMS 1-3 piezocomposite as the flexible active material, alongside an Ag/PDMS composite layer functioning as the flexible electrical conductor. The resulting transducer exhibited a center frequency of 2 MHz and a bandwidth of approximately ~50%. Similarly, Peng et al. [20] reported a flexible transducer operating at a center frequency of 5 MHz. The 1-3 piezocomposites usually exhibit reduced acoustic impedance compared to PZT film, enabling the creation of a wide bandwidth transducer without the need for an additional acoustic matching layer. This particular transducer demonstrated remarkable flexibility and a wide bandwidth of approximately 47%.

2) Role in Biomedical Applications and Suitability: Flexible single-element piezoelectric transducers offer a practical solution for biomedical imaging applications, especially in the context of photoacoustic and thermoacoustic imaging. Their simplicity and flexibility make them well-suited for these tasks. They can conform to the body's contours comfortably, ensuring effective signal acquisition. The wide bandwidth of these transducers allows them to capture a broad range of photoacoustic or thermoacoustic signals, which is advantageous for differentiating tissue types and improving imaging depth. However, these transducers have some limitations. One key disadvantage is their lower image resolution. For this reason, they may not be the first choice for applications demanding high-resolution imaging. Nevertheless, they are cost-effective and straightforward, offering a balance between performance and simplicity. When choosing this type of transducer for imaging, the tradeoff between resolution and cost-effectiveness is a critical consideration.

In the realm of biomedical sensing, flexible single-element piezoelectric transducers can play a valuable role. Their comfort and conformability to the skin make them suitable for applications such as blood flow measurement, blood pressure monitoring, and heart rate measurement. These sensors can be integrated into wearable and implantable devices, ensuring noninvasive and user-friendly solutions for long-term monitoring. Despite their advantages in terms of comfort and noninvasiveness, these transducers may face challenges in achieving high accuracy, particularly during strenuous activities. The choice of this type of transducer for sensing applications often revolves around the tradeoff between comfort, invasiveness, and the specific monitoring needs of each application.

Flexible single-element piezoelectric transducers can also find application in biomedical therapy. These transducers can be integrated into wearable therapeutic devices to provide focused treatment. Their comfort and adaptability to various body parts make them suitable for therapeutic applications, ensuring patient compliance and comfort. Nevertheless, some limitations need to be considered. Achieving high precision, especially in deformity measurements, can be challenging. When selecting this type of transducer for therapy, the choice often depends on the specific therapeutic goals, comfort, and ease of use, balanced with the need for precision and effectiveness.

B. Flexible Micromachined Array

1) Sparse Array With Conventional Bulk Piezoelectric Transducers:

a) Structure and design of sparse array: To fabricate a flexible array, one approach was to fabricate conventional PUTs with bulk materials. After adding flexible interconnections, the array was packaged with flexible substrates. Yang et al. [76] fabricated a flexible transducer 4×4 array, each with a size of 1 mm on a polyamide base with PDMS packaging. Hu et al. [16] developed a 2-D matrix flexible array ultrasonic (US) transducer that is capable of conforming to complex surfaces. This transducer is composed of a 10×10 array of units, each of which is made up of rigid PZT/epoxy 1-3 piezocomposites as the active elements of the transducer; 20-μm-thick multilayered serpentine copper traces were applied as the electrical conductors, and the array was encapsulated in Ecoflex elastomer membranes. Xue et al. [98] developed a 4×4 sparse array with PZT-5A rigid bulk materials and packed with flexible PDMS at 10.6 MHz with averaged bandwidth of -61%. Liu et al. [99] reported a 3×3 array with PZT materials, which was packaged with PI. Although the transducer units remained rigid, the utilization of serpentine electrical interconnects in the design imparted flexibility to the 2-D arrays. Yet, the design of the interconnection will be complicated by the increase in the element numbers, which limited the density of element numbers.

b) Role in biomedical applications and suitability: Sparse arrays with conventional bulk piezoelectric transducers offer a reliable and cost-effective solution for biomedical applications where rigidity is not a constraint. They are suitable for

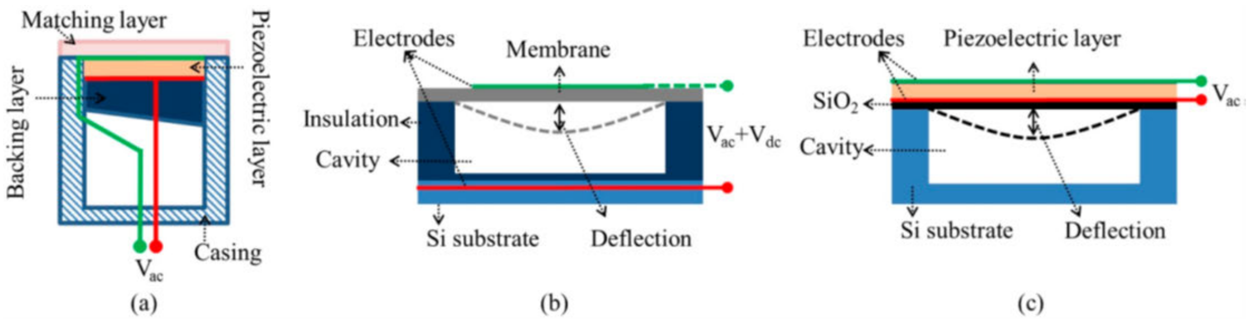

Fig. 2. Structure of typical ultrasound transducers. (a) Bulk piezoelectric transducers. (b) PMUTs. (c) CMUTs. Adapted from [95].

applications where stable and rigid transducers can provide accurate results. However, their limited flexibility can be a disadvantage in applications requiring patient comfort, body conformity, or extended wear times. The choice of these transducers depends on the specific application's needs, weighing their advantages in terms of reliability and performance against the disadvantages related to flexibility and comfort.

2) Capacitive Micromachined Ultrasound Transducers:

a) Structure and design of CMUTs: CMUTs have received significant attention due to their potential for use in medical imaging. These transducers have several advantages over traditional piezoelectric materials, such as wide bandwidth, ease of miniaturization, and the ability to be fabricated on a large scale [78]. A CMUT is a miniaturized plate capacitor that consists of three main parts: a flexible plate electrode suspended and bent out-of-plane at the top, a gap between the two electrodes enabling flexible electrode bending and vibration, and a stationary electrode positioned at the bottom [100]. Upon applying an alternating current (ac) stimulus across the two electrodes, the upper flexible electrode undergoes mechanical vibrations due to electrostatic forces, resulting in the generation of ultrasound waves. Conversely, US waves induce vibrations of the flexible top electrode, resulting in changes to the CMUT's capacitance. With a relatively large bandwidth, the flexibility of CMUT arrays and devices enables a wide range of applications in health monitoring and medical imaging [101].

There are currently two strategies for creating flexible CMUT arrays: integrating miniaturized CMUTs as "rigid islands" on or into a flexible substrate with stretchable or flexible interconnects or building the entire CMUTs from intrinsic stretchable or flexible materials. The development of flexible substrates and stretchable conductors has greatly improved the flexibility and performance of CMUTs. Caronti et al. [102] successfully produced a flexible CMUT array through trench refilling with PDMS, featuring a center frequency of 11 MHz and a −6-dB bandwidth spanning 100%. Pang and Chang [36] introduced a flexible CMUT array featuring a center frequency of 880 kHz. By comprising an oxide-PET substrate and SU-8 sidewall, the array was transparent with good flexibility [36]. Zhuang et al. [103] reported a circular CMUT array based on silicon and PDMS interconnections with a center frequency of 4.5 MHz, which was wrapped on a catheter with a diameter of 0.9 mm for imaging applications, specifically intravascular ultrasound (IVUS). Recently, Omidvar et al. [104] reported a 4-MHz CMUT with −6-dB bandwidth over 75%. Each element comprised three layers: PI substrate, SU-8, and parylene-C with Young's modulus of 3.5, 6.5, and 4.5 GPa, respectively. The array was characterized under a radius of curvature of 3 cm [104]. Yet, the flexible CMUT suffered from the limitations of electrical stability due to the dc bias and relatively low-pressure output.

b) Role in biomedical applications and suitability: CMUTs are highly versatile transducers with significant utility in various biomedical applications. Their wide bandwidth, adaptability to body contours, and potential for miniaturization render them ideal for applications in medical imaging, such as photoacoustic and thermoacoustic imaging, where high-resolution images are imperative. In addition, they prove to be well-suited for tasks such as blood flow measurement, blood pressure monitoring, heart rate measurement, and muscle activity monitoring, enabling precise measurements while ensuring patient comfort. In therapeutic applications such as bone injury treatment and wound management, CMUTs offer focused treatment and adaptability to different body parts. Nevertheless, it is essential to acknowledge that the fabrication of CMUTs necessitates intricate microfabrication and nanofabrication techniques. This complexity, in turn, contributes to a relatively intricate and potentially costly manufacturing process. Furthermore, CMUTs demand relatively high operating voltages, which can result in elevated power consumption, particularly at higher frequencies.

3) Piezoelectric Micromachined Ultrasound Transducers:

a) Structure and design of PMUTs: PMUTs are a type of transducer that has been implemented in a variety of applications such as fingerprint sensors, rangefinders, and intravascular US. These transducers generate US waves through the thickness oscillations of a piezoelectric layer [95]. PMUTs operate based on flexural oscillations of an excited piezoelectric membrane. A PMUT is composed of a piezoelectric membrane that is enclosed between two slender electrodes and suspended to vibrate in a flexural mode. Upon the application of an electric field through the membrane's thickness, lateral strains are generated in the planar orientation, resulting in the bending of the membrane. The PMUT's resonant frequency is influenced by factors such as form factors, stress conditions, and the mechanical properties of the membrane. For flexible PMUTs, Lee et al. [105] reported a 16-element linear array with silicon on insulator (SOI) wafer and

PDMS substrate. Although the device showed its flexibility over a curvature of 5 mm, the electrical impedance was high (>1 kΩ) for each element with a center frequency of 690 kHz. Liu et al. presented a PVDF PMUT on PI and Kapton base, which gave a frequency of 198 kHz [42] and a 3 × 3 array with PVDF on PET substrate with a center frequency of 160 kHz [53]. Jeong [41] reported a flexible 10-kHz PMUT with P(VDF-TrFE) and PI at a frequency of 10 kHz. However, current flexible PMUT transducers are mostly limited to a working frequency lower than megahertz, which is not suitable for biomedical applications.

b) Role in biomedical applications and suitability: PMUTs have been implemented in various applications, including fingerprint sensors, rangefinders, and IVUS. These transducers generate ultrasound waves through the flexural oscillations of a piezoelectric layer, making them suitable for specific imaging and sensing applications. While they offer flexibility and adaptability, their operational frequencies are typically limited to lower ranges, making them less suitable for high-frequency imaging. PMUTs find their niche in applications where lower frequencies are acceptable and where their adaptability, miniaturization potential, and ease of integration into flexible substrates are advantageous. Their advantages include flexibility and ease of integration into flexible substrates, making them suitable for applications with specific frequency requirements and where conformability to body contours is essential.

C. Flexible Laser-Generated Ultrasound Transmitters

1) Structure and Design of Flexible LGUS Transmitters: Inspired by the mechanism of photoacoustic, which describes the formation of sound waves following pulse or burst light absorption in a material sample [106], laser-generated ultrasound (LGUS) transmitters have been reported to transmit the laser energy into short acoustic pulses with high power density, high frequency, and broad bandwidth [107]. A laser ultrasound transducer typically comprises a laser light absorption layer with a strong light absorption ratio, along with a thermal expansion layer featuring a notable thermal coefficient of volume expansion [108]. For the thermal absorption layer, to obtain high-performance laser ultrasound transducers, active research has been focused on absorbing materials, such as carbon black, CNTs [109], graphite, metallic films [110], gold nanostructures [107], AuNP [111], and candle-soot nanoparticles (CSNPs) [112], [113].

Although most LGUS transmitters were mounted on rigid substrates such as glasses [68], [114], [115], recent works have reported flexible LGUS transmitters based on the PDMS or PI substrates [108]. These transmitters harness the photoacoustic mechanism and are designed to generate short acoustic pulses by converting laser energy. Notably, they emphasize flexibility or stretchability in their design, making them suitable for applications where conformal adhesion is crucial. An example of this approach is seen in [108], where a composite of CSNPs and PDMS is utilized. This composite demonstrates high-frequency and high-amplitude pressure outputs while also maintaining stretchability. These advancements open pathways for creating flexible ultrasound devices suitable for both diagnostic and therapeutic applications.

Mounted on optical fibers, the LGUS transmitters also showed their potential for intravascular flexible devices [116], [117]. Compared with traditional piezoelectric transducers, the LGUS transmitters have a less complicated structure for flexible design and fabrication. Although can generate ultrasound with a noncontact method, the LGUS transmitters still have safety issues and require additional devices for receiving ultrasound signals.

2) Role in Biomedical Applications and Suitability: In the realm of biomedical applications, flexible LGUS transmitters hold promise for various purposes. Their ability to generate short, high-frequency acoustic pulses makes them suitable for diagnostic applications such as photoacoustic and thermoacoustic imaging. By converting laser energy into ultrasound waves, LGUS transmitters offer a noninvasive and precise imaging modality for visualizing tissues, detecting abnormalities, and monitoring disease progression. Moreover, their flexibility and stretchability, especially when incorporated into flexible substrates, enable conformal adhesion, which is critical for applications involving complex anatomical surfaces or intravascular devices. LGUS transmitters also demonstrate potential in therapeutic applications such as localized drug delivery and neuromodulation, where precise targeting and minimally invasive approaches are paramount. Despite their advantages, LGUS transmitters require additional devices for receiving ultrasound signals, and safety concerns related to laser use must be addressed. Overall, their versatility and suitability for both diagnostic and therapeutic applications make flexible LGUS transmitters a valuable technology in the biomedical field.

D. Wearable and Integrated Ultrasound Systems

1) Structure and Design of Wearable and Integrated US Systems: The realm of US technology is undergoing a remarkable transformation. US systems are no longer confined to clinical settings but are becoming increasingly wearable, integrated, and accessible. This section offers an in-depth exploration of these innovations, tracking the evolving trends that have reshaped ultrasound devices. We examine the progression from traditional stationary systems to contemporary wearable and integrated solutions, emphasizing the importance of user-friendliness and convenience.

The evolution of ultrasound technology is driven by several significant trends, underlining its adaptability and accessibility. First, there is a noticeable shift from conventional stationary ultrasound systems to wearable [118], [119], [120], [121], [122], [123], [124], [125] and integrated solutions [126], [127]. This transition responds to the growing demand for point-of-care ultrasound applications and the need to provide personalized healthcare in various environments. Modern wearable ultrasound devices, characterized by their portability and lightweight design, offer considerable advantages over their bulkier predecessors. This shift toward wearable technology in the healthcare sector reflects a broader societal change toward health monitoring and personalized medicine. In essence, ultrasound technology is no longer tied to traditional clinical settings but is now more versatile and dynamic.

As ultrasound systems evolve, so does the domain of communication technologies. A significant transformation has occurred in how data are transmitted and received. In the past, ultrasound machines were linked by numerous wires and cables, restricting both device and operator mobility. However, with the adoption of wireless communication technologies, such as Bluetooth [121] and Wi-Fi [124], [126], data exchange between ultrasound devices and external platforms, such as smartphones [121], [126], tablets, or cloud-based servers, has become seamless. This enhances flexibility and mobility for clinicians and users during ultrasound examinations. It streamlines workflows and makes ultrasound technology more accessible and user-friendly. A noteworthy advancement in the field of ultrasound technology is the shift from traditional control and data acquisition methods to innovative approaches integrated into wearable smart systems. Previously, ultrasound systems depended on robust, stationary control units that often interfaced with large workstations or PCs. However, wearable smart devices, such as cellphones and smartwatches, have transformed this landscape. These compact yet powerful devices now function as comprehensive control units and data acquisition platforms for ultrasound examinations.

Integrating various components into a coherent wearable ultrasound system is a complex yet critical aspect of modern ultrasound technology. These components encompass ultrasound transducers, electronic circuitry, power sources, and form factors. The integration process entails careful consideration of factors such as form factor design, power management, data processing, and user interface. These elements are harmonized to create a user-friendly and efficient ultrasound system. In the following, various biomedical applications of these wearable or integrated ultrasound systems are explored, further emphasizing their crucial role in the realm of flexible ultrasound transducers. A summary of several wearable and integrated US systems can be found in Table III.

2) Role in Biomedical Applications and Suitability: Wearable and integrated ultrasound systems have significantly impacted various biomedical applications, enhancing the way healthcare is delivered and diagnostic information is obtained. In the realm of medical imaging, they provide a versatile tool for point-of-care diagnostics, facilitating imaging techniques such as photoacoustic imaging, thermoacoustic imaging, and conventional medical and tissue imaging. These systems offer improved accessibility and the ability to adapt to complex clinical scenarios. In biomedical sensing, they enable applications such as blood flow measurement, blood pressure monitoring, and heart rate measurement. Their wireless capabilities and integration into smart devices enhance data acquisition and user experience. Finally, in the field of biomedical therapy, they have applications in bone injury treatment, chronic wound healing, drug delivery, and neuromodulation. The flexibility and adaptability of these systems make them valuable tools for delivering targeted therapies and interventions. Overall, wearable and integrated ultrasound systems are instrumental in transforming the landscape of healthcare by providing accessibility, flexibility, and user-friendliness to both clinicians and patients.

IV. Flexible Ultrasound Transducers in Biomedical Applications

In recent years, flexible ultrasonic transducers have become increasingly popular for applications involving imaging, sensing, and therapy, especially those that involve curved surfaces. A flexible transducer minimizes the burden of wear and conforms closely to surface contours, overcoming the limitations of traditional rigid transducers. Flexible ultrasonic transducers are useful for a broad spectrum of applications because they are versatile and can provide comparable measurement and imaging results. This section offers an overview of the various kinds of flexible US transducers documented in the existing literature.

A. Biomedical Imaging

In this section, we discuss the applications of flexible US transducers in biomedical imaging. A summary of several properties of flexible US transducers can be found in Table IV. Fig. 3 illustrates various biomedical imaging applications incorporated with flexible US transducers.

1) Harnessing Light and Sound: Photoacoustic Imaging: The photoacoustic imaging technique is a newly developed, noninvasive imaging technique that provides optical contrast to targets deep within tissues. It can be used to image endogenous and exogenous chromophores that are optically absorbing and have applications in imaging cancers, cardiovascular diseases, and others [128]. Flexible transducers work well in photoacoustic imaging applications, such as tomography arrays that require conformal or curved transducers [77], [129], [130]. Roy et al. [129] developed an integrated imaging system that combines ultrasound and photoacoustic modalities, utilizing PZT-based flexible bulk transducer arrays for enhanced functionality. The transducer having 50 elements was fabricated using bulk PZT-5H material and employed the usage of a flexible epoxy-based matching layer. An evaluation of the biomedical imaging performance of the device was carried out by attaching it to the Vantage 256 Research US System (Verasonics Inc.) for obtaining B-mode images of custom-made phantoms that contained metal targets. Furthermore, it served as a receiver for acquiring the A-line photoacoustic signal. It was found that the peak voltage obtained at 0-dB gain was 22 mV, and the fractional bandwidth at -6 dB was 37% [129]. Ghavami et al. designed the first flexible transparent CMUT 64-element array for through-illumination photoacoustic tomography applications [see Fig. 3(a)]. An adhesive wafer bonding technique was utilized to fabricate the device. Benzocyclobutene (BCB), ITO, silicon nitride, and PDMS were used as the adhesive, electrodes, membrane, and flexible backing, respectively. The device demonstrated 67% transparency in the visible range as a result of the adhesive wafer bonding technique. The fabricated array was bent to a radius of curvature of less than 5 mm without affecting the PDMS connections. The reliability was

TABLE III
WEARABLE AND INTEGRATED ULTRASOUND SYSTEMS: KEY FACTORS AND CHARACTERISTICS

Application	Transducer types	Wearable or Integrated system	Center frequency [MHz]	Power source	Communication	Ref
Cardiopulmonary Monitoring	single element	Wearable	1 MHz	Integrated circuits (3.3V) Transducer stimulation (5∼20V)	Wired	[118]
Muscle States Classification	single element	Wearable	3.5 MHz	2.5 W	Wired	[119]
Gesture Recognition	four-channel single element	Wearable	5 MHz	Driving pulse (40V)	Wired	[120]
Mental Health Treatment	four-channel single element	Wearable	1.5 MHz	N/A	Bluetooth	[121]
Prosthetic Control	eight-channel single element	Wearable	5 MHz	Battery (2V 3Ah)	Wired	[122]
Gesture Recognition	four-channel single element	Wearable	5 MHz	Driving pulse (40V)	Wired	[123]
Muscular Monitoring	eight-channel single element	Wearable	5 MHz	Battery (2V 3Ah)	Wired Wi-Fi	[124]
Muscle Force Estimation	single element	Wearable	5 MHz	Ultrasound transducer (70V) electric components (12V)	Wired	[125]
Muscular Monitoring	256 elements array	Intgrated	4 MHz	Battery (614 mW)	Wi-Fi	[126]

tested over 500 cycles of bending to a radius of 2 cm without the PDMS connections breaking. A photoacoustic imaging experiment with a 100-mm-diameter aluminum wire target was conducted to evaluate the lateral and axial resolutions of the fabricated CMUT arrays, respectively, with a signal-to-noise ratio (SNR) of 46 dB. Further demonstrations of photoacoustic imaging were conducted on a chicken breast sample that was inserted into a tube that contained bovine blood in the form of a tissue-mimicking phantom [130]. Liu et al. [77] fabricated a single-element US sensor with a sandwich structure consisting of 150-nm-thick amorphous ITO on both sides of a 110-μm commercial PVDF film (Piezo Film Sheets, TE connectivity). Various chords of curvature between 10 and 30 mm were used to evaluate the flexibility of the transducer. To determine the transducer's performance, an A-Scan result was obtained from a customized phantom containing four alphabet letters with a diameter of 1 mm, which were positioned on an inclined plane in order to simulate the actual conditions of wearables. Following the processing of the acoustic signals, a 3-D image was obtained. There was a close match between the distance from the surface to the letter and the design parameters [77].

2) From Heat to Sound: Thermoacoustic Imaging: Thermoacoustic imaging (TI) refers to imaging that combines microwave absorption with US technology. It has the potential to serve as an alternative imaging technique in the realm of biomedical applications [134], [135], [136], [137], [138]. In TI applications, traditional linear US probes have disadvantages, such as the need for information about the scanning radius to be gathered from a variety of elements. Furthermore, in the case of irregular samples, the scanning radius must be large enough to cover the longest axis of the imaging geometry [131]. The flexible US transducer, on the other hand, can be brought into close contact with samples of any shape. A number of commercially available flexible transducers have been introduced into TI systems [131], [132]. Ji et al. [131] proposed the sample-cling-scanning (SCS) model that combined flexible multielement transducers and adaptive back projection algorithms to acquire microwave signals [see Fig. 3(b)]. The transducer used here was a commercial 64-element transducer (7.5S64-0.5*10, Doppler Ltd., China), which has the smallest curvature radius of approximately 5 mm and can endure at least one hundred thousand bending cycles. The main advantage of the SCS model was its capability to adapt to samples of any shape without the requirement of a coupling liquid, and the acquired information can be used to reconstruct images by an adaptive back projection algorithm. Furthermore, the energy transmission efficiency of SCS was approximately 80% higher than that of conventional circular scanning (CS) mode. To demonstrate the quality of the proposed systems' TI, a tumor imaging experiment was conducted. This results in TI that was similar to X-ray images but with a higher degree of contrast [131]. Xiang et al. [132] measured the TI signal of rabbit liver using a commercial 96-element flexible US transducer (Japan Probe Company Ltd.). TI acquired in vivo was consistent with the liver's anatomical structure [see Fig. 3(c)] [132].

3) Peering Into the Body: Medical and Tissue Imaging: It is imperative that flexible US transducers are developed for biomedical imaging applications. It is possible to predict many diseases and disorders through the monitoring of various physiological functions and the acquisition of medical imaging within the human body [128]. Typical biomedical imaging applications using flexible transducers include needle guidance [139], localization imaging [133], 3-D imaging [16], and medical imaging [16], [39], [93], [140]. Chen et al. [133] introduced an innovative one-layer island bridge configuration featuring a 2-D flexible ultrasonic RC array. This design

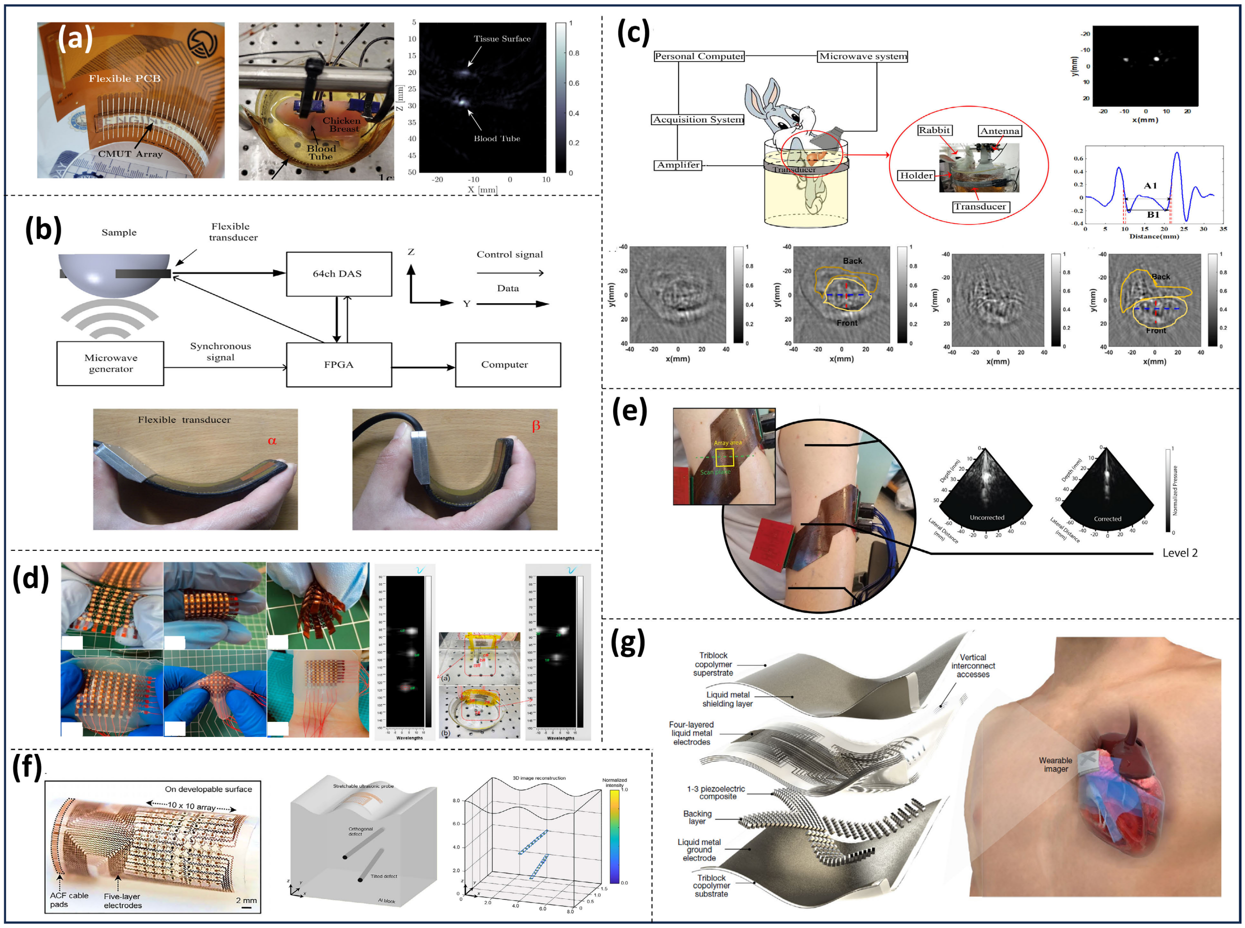

Fig. 3. Flexible transducer for photoacoustic, thermoacoustic, and biomedical imaging. (a) Ultrasound transducer type-CMUT arrays. Photographs of the fabricated flexible transparent CMUT arrays with wire bonded to a flexible PCB with a cut-out in the back of the array (left). Schematic of ex vivo photoacoustic tomography: chicken breast phantom with an embedded tube filled with bovine blood (middle) and photoacoustic image of the blood tube acquired by the fabricated CMUT array (right). Adapted from [130]. (b) Ultrasound transducer type-piezoelectric-polymer matrix composites arrays. The experimental setup of shape-adapting TA imaging system (top). The photographs of the flexible multielement transducer with two different shapes (bottom). Adapted from [131]. (c) Experimental setup of the imaging system (top). In vivo liver imaging with thermoacoustic imaging (bottom). Adapted from [132]. (d) Ultrasound transducer type-rigid piezoelectric materials with flexible interconnect. The photographs of the flexible array (top left). The experimental setup of the imaging platform (lower left). The imaging results (right). Adapted from [133]. (e) Ultrasound transducer type-rigid piezoelectric materials with flexible interconnect. Human humerus imaging using the FlexArray. Adapted from [39]. (f) Ultrasound transducer type-sparse array with rigid piezoelectric transducers, flexible interconnect, and substrate. Diagram of the stretchable ultrasonic transducer array (left), and experimental setup schematics and the reconstructed 3-D image (right). Adapted from [16]. (g) Ultrasound transducer type-rigid piezoelectric materials with flexible interconnect. Wearable cardiac imager: exploded view schematics highlighting key components (left) and operational principle (right) of the wearable imager. Adapted from [93].

effectively mitigates the ringing effect (excessive vibration) while enhancing axial resolution [see Fig. 3(d)]. Adaptable flexible electrodes were designed in accordance with the row-column addressing principle. As a part of the flexibility and reliability testing, the array and encapsulator were stretched, twisted, and bent in order to verify their mechanical properties. An experiment involving multimodal localization imaging was performed utilizing the Verasonics system (Verasonics Inc.), targeting both multitarget steel pins positioned on curved and planar surfaces. The achievement of autonomous and organized excitation of individual units using 16 leads was facilitated by the innovative N + N flexible electrode configuration. This design enables the array to reconstruct target shapes by combining multisection images. Utilizing a 3-D scanner and the array, the experiment involved acquiring 2-D planar scan images using three randomly positioned steel pins in the sink. The results indicated that the measured target distances closely matched the theoretical values, exhibiting accuracy errors of 1.54%, 3.27%, and 4.14% for the respective pins [133]. Culjat et al. [139] presented a study detailing the creation of a conformal and flexible ultrasound transducer. This transducer utilized bulk PZT elements positioned on etched silicon islands and linked together through a patterned flexible PI joint. Transducers were bent to radii of curvature smaller than 1 cm in order to demonstrate their high flexibility. After 10 000 bending cycles, the PI/parylene joints did not exhibit visible damage or measurable electrical degradation. The transducer's functionality was verified by inserting a target needle into a soft tissue phantom. Through the use of a space–time backward propagation image reconstruction algorithm, a composite image clearly illustrating the needle target was reconstructed, and an SNR of 39 dB was achieved from a needle target [139]. Elloian et al. [39] developed a 256-element flexible 2-D ultrasound piezoelectric transducer array with geometric phase correction [see Fig. 3(e)]. Throughout various bending stages, the 2-D array upholds favorable performance metrics, including a high SNR and minimal elemental crosstalk. To enhance image reconstruction during flexible transducer deformation, a unique geometric phasing algorithm was formulated. This algorithm applies imaging corrections utilizing established radii of curvature in the x- and y-axes. In vivo, an image of the human humerus' surface curvature was obtained by directly wrapping the device around the arm of a healthy male [39]. Mierzwa et al. [140] presented 16-element flexible devices that can be operated in flat, concave, or convex configurations. Branched-vessel vascular phantoms that simulate the human vasculature were used to test point-of-care imaging and patient monitoring capabilities. For comparison, a standard 64-element linear array probe (Cephasonics TL071) was also used to image the phantom. Consequently, high accuracy was achieved when imaging vessels of 4 and 6 mm diameters, with percent errors of 2.5% and 3.3%, respectively. Compared to the larger commercial linear array, the flexible devices had percent errors of 2.5% and 1.7%, respectively, for the same vessels of 4 and 6 mm [140]. Based on an island-bridge layout with multilayer electrodes, Hu et al. [16] fabricated arrays containing piezoelectric 1-3 composites encapsulated in thin and compliant silicone elastomers [see Fig. 3(f)]. During flexibility testing, it exhibited more than 50% stretchability. Multiview imaging of a complex customized aluminum workpiece with embedded defects (2 mm in diameter, orthogonal to the side surface) has been achieved. For the reconstruction of the multiview images, the synthetic aperture focus (SAF) method was applied. SAF is advantageous in that it allows sparse transmitter–receiver schemes, thus minimizing the number of simultaneously active elements while maintaining image quality [16].

Hu et al. [93] engineered a wearable ultrasonic gadget designed to perform continuous and real-time assessment of cardiac function [see Fig. 3(g)]. The innovation lies in enhanced device-skin mechanical coupling through novel design and material fabrication. The device incorporates arrays of piezoelectric transducers, liquid metal composite electrodes, and encapsulation using triblock copolymers. It utilizes anisotropic 1-3 piezoelectric composites along with eutectic gallium–indium liquid metal to create high-density stretchable electrodes. A customized algorithm controls the activation sequence of 88 channels to reconstruct sectorial images sequentially, resulting in optimized frame rates for specific echocardiographic views. Moreover, an advanced deep learning model automatically derives cardiac chamber dimensions from continuous image recordings, enhancing the potential of biomedical imaging techniques. Lin et al. [126] have made notable strides in wearable ultrasound technology, addressing challenges of wire connections, motion tracking, and data interpretation. Their innovative advancement entails an integrated and self-sustained ultrasonic-system-on-patch (USoP), combining a compact and flexible control circuit, an array of ultrasound transducers, and machine learning algorithms for tracking moving targets [126]. This innovation enables continuous monitoring of deep tissue physiological signals, such as central blood pressure, heart rate, and cardiac output, for extended periods. The USoP utilizes a gel-free acoustic coupling interface, enhancing resolution, noise reduction, and durability in tissue sensing. Multimodal tissue imaging, encompassing A-mode, B-mode, and M-mode, further enriches its capabilities. Employing domain adaptation techniques, machine learning networks have been optimized for generalization across diverse image datasets. The USoP demonstrates the promising potential for real-time, hands-free monitoring of deep tissue signals during various human activities, serving the burgeoning field of the Internet of Medical Things.

B. Biomedical Sensing

In this section, we summarized recent research progress on customized flexible transducers in different health monitoring and sensing applications. Table V summarizes the key characteristics of the fabricated flexible transducers. Detailed illustrations of the sensors and experiment configuration can be found in Fig. 4.

1) Vascular Vital Signs: Blood Flow and Pressure Monitoring: Cardiovascular conditions are clearly reflected in variations in blood vessels. Therefore, continuous monitoring and operator-independent measurements are becoming increasingly important for patients' health and early diagnosis and prognosis. Cannata et al. [141] fabricated a flexible and low-power consumption system to measure the blood flow rate [see Fig. 4(a)]. A slab transducer electrode pattern and two diffraction-grating transducers were fabricated on flexible piezoelectric-polymer films. The peak blood velocity measurement through the rabbit infrarenal aorta was within 1.7% of that obtained with the calibrated HD11XE duplex US imaging system (Philips Healthcare, Bothell, WA, USA) [141]. Wang et al. [142] developed a wearable and lightweight Doppler US device for continuously monitoring blood flow through arteries [see Fig. 4(b)]. A 3 $\times$ 3.1-3 composite piezoelectric transducer array was encapsulated and cured into a soft substrate using Ecoflex. With the device attached to the curved skin surface, it is possible to determine the blood flow velocity with no coupling agent, low pressure, and no imaging experience from the operator. This device demonstrates the potential for long-time monitoring of blow flow velocity [142]. A piezoelectric 1-3 composite ultrasonic sensor was developed by Peng et al. [20] to continuously monitor a patient's blood pressure. It is composed of PZT-5A microscopic rods embedded in a PDMS polymer matrix. The piezoelectric 1-3 composite was coated with flexible electrodes on both surfaces. The reported ultrasonic sensor was capable of detecting a variation in arterial diameter of 0.49 mm. As a result, blood pressure can be estimated with an error of $\pm$0.87 mmHg, which is consistent with the commercial blood pressure monitor [20]. A conformal and stretchable ultrasonic device to continuously monitor the waveform of central blood pressure was demonstrated by Wang et al. [17]. The device was constructed from anisotropic 1-3 composites and soft structures. The device is only three orders of magnitude thinner than conventional US probes while still delivering comparable performances. It can be stretched and twisted to provide conformal contact with the human skin while

TABLE IV
FLEXIBLE ULTRASONIC TRANSDUCER FOR BIOMEDICAL IMAGING

Applications	Number of Elements	Center Frequency [MHz]	Thickness [mm]	Size	Resolution [um]	Bandwidth @-6 dB	Ref
Photoacoustic imaging system	50	2.73	N/A	0.39 mm×5 mm per element	N/A	37 %	[129]
Photoacoustic imaging system	64	3.5	N/A	0.415 mm× 7 mm per element	lateral: 293 axial: 382	80 %	[130]
Photoacoustic imaging system	single element	6.7	N/A	2 cm × 2 cm	lateral: 400 axial: 272.5	86.3 % @-3 dB	[77]
Thermoacoustic imaging	96	2.5	N/A	10 mm × 3 mm per element	N/A	82 %	[132]
Thermoacoustic imaging	64	7.5	N/A	0.5 mm × 10 mm per element	N/A	70 %	[131]
Localization imaging	64	1.95	N/A	side length of 0.9 mm per element	N/A	N/A	[133]
Needle guidance	8	12	N/A	1.8 mm × 1.8 mm per element	N/A	2 MHz @ -3 dB	[139]
Medical imaging	256	1.4	1.26	N/A	axial: 2000	41 %	[39]
Medical imaging	16	5/7	N/A	N/A	N/A	N/A	[140]
3D imaging	100	3.5	N/A	1.2 mm × 1.2 mm per element	lateral: 344∼789 axial: 610	47.11%	[16]
Cardiac imaging	88	3	1	12.7 mm 32 elements aperture	lateral: 1800∼3000 axial: 400	55%	[93]
Deep tissues imaging	112 (2 MHz) 256 (4 MHz) 32 (6 MHz)	2, 4, 6	N/A	25.4 mm (4 MHz Patch size)	axial 604 (2 MHz) 333 (4 MHz) 229 (6 MHz)	∼50%	[126]

maintaining excellent beam directivity and penetration depths up to 40 mm, which is essential for clinical use. As a result of its mechanical compliance and lightweight design, this device supports a gel-free mode of operation. Transducers and skin can be intimately connected by a thin layer of sticky silicone, which eliminates the pressure when applying a traditional tonometer. With a relative measurement uncertainty of 1% and a precision of within 2 mmHg, the ultrasonic device outperformed a commercial tonometer. In addition, the measurement is independent of the operator's skill level. With this device, the amplification effect, which is hard to observe in the current clinical setting, was demonstrated via a direct diameter measurement approach from multiple body parts. Overall, the proposed device showed promising clinical potential using advanced microfabrication techniques and strategic materials' integration [17].

2) Clinical Insights: Medical Sensing and Vital Sign Monitoring: Due to the complicated contours of the human body, flexible and wearable sensors are ideal for whole-body biomedical sensing. Furthermore, flexible sensors can be customized to fit different applications and are smaller than rigid systems currently available on the market. Hamelmann et al. [143] integrated 37 piezoelectric transducer elements into a flexible PDMS substrate [see Fig. 4(e)]. With the proposed dynamic apodization adaption method, the flexible transducer array can be wrapped around the abdomen and continuously measure the fetal heart rate without the need to manually reposition the transducer. The total acoustic energy transmitted to the fetus was also minimized for safety reasons [143]. Wang et al. [144] presented a flexible and real-time photoacoustic imaging device, consisting of flexible optical fiber bundles and a commercial 128-element flexible transducer (10S128-5.0*5, Doppler Ltd.) to monitor the physiological changes of breast tumors. In addition to its wide field of view, the system is reported to have an imaging resolution of 130 μm. This device can adapt to curved surfaces with a radius of as little as 5 mm, making it suitable for morphological adaptation in imaging applications [144]. According to Xue et al. [98], [145], they have developed a low-cost and flexible ultrasonic transducer for monitoring human muscle activity at different depths using PZT-5A and PDMS as the substrate [see Fig. 4(c)]. Compared with surface electromyography measurements, the transducer produced a higher SNR and avoided muscle crosstalk. In addition, muscle fatigue did not degrade the function of the US transducer. It is also possible to customize and adapt the fabricated transducer to fit different geometric shapes of the target surface [98], [145]. Shea et al. [146] recently reported using a low-density flexible array US transducer to assess spinal deformity by measuring the spinous process location and the spinal deformity angle (also called the Cobb angle) [see Fig. 4(d)]. Embedded in thermal deformable plastic plates, this custom-designed 1-D array achieved a minimum radius of curvature of about 30 mm. Therefore, it conforms to the complicated back contour of a patient and minimizes air gaps between the transducer and the body. Using a linear motion stage, 3-D US images can be reconstructed by sliding the array over the spine phantoms. Four different configurations of transducer arrays were tested in terms of measurement accuracy and intensity distribution. A linear array resulted in the best measurement with an error of 2.7 mm $\pm$ 1.5 mm, while an S-shaped transducer achieved the worst measurement with an error of 4.8 mm $\pm$ 3.2 mm due to the asymmetrical distortion [146].

C. Biomedical Therapy

In this section, we provide an overview of the recent advancements in customized flexible transducers for use in treatment and neuromodulation applications. Table VI highlights the key features of the manufactured flexible transducers, while Fig. 5 provides a detailed illustration of the sensors and experimental setup used in these studies.

Ultrasound is a popular tool in biomedical treatment due to its many advantages, including noninvasiveness, excellent tissue penetration, and targeted energy delivery. High-intensity focused ultrasound (HIFU) is a promising technique for therapeutic applications that utilizes high energy to damage targeted tissues. The damage can be either thermal or mechanical, depending on the duty cycle of the HIFU. The thermal effect is achieved by using a high-duty cycle, while a low-duty cycle produces a mechanical effect. Thermal effects rely on energy absorption, resulting in heating and coagulation of tissue, while mechanical effects depend on cavitation, creation, and collapse of gas bubbles in tissues. Low-intensity pulsed ultrasound (LIPU) delivers lower energy to tissues, stimulating cellular processes through subtle mechanical vibrations [147], [148], [149]. These vibrations promote tissue regeneration, angiogenesis, and the production of growth factors. Flexible ultrasound transducers offer a patient-friendly solution for noninvasive procedures, which makes them a suitable option for long-term treatments. In-home treatment using flexible ultrasound devices can be a cost-effective solution, reducing the burden on healthcare systems and improving accessibility to patients. Although flexible devices can be challenging to fabricate, the benefits they offer justify the efforts in developing them. The noninvasive nature of ultrasound treatments also leads to fewer complications and shorter recovery times for patients, making it an effective tool for biomedical treatment.

1) Mending and Nurturing: Injury Treatment and Deep Brain Stimulation: Liu et al. [150] demonstrated the development of an arrayed flexible PMUT with a sandwich structure [see Fig. 5(a)] [150]. The flexible substrate of PDMS was integrated with a bulk lead PZT array and flexible metal line interconnection. The flexible interconnection possesses a tensile strain capacity of 61.5%, enabling the PMUT to stretch by 25%, meeting the requirements for biological deformation. This flexibility was evaluated through an in vitro pork tissue test, showing successful ultrasonic wave propagation through various types of tissue. These findings suggest potential applications of the flexible PMUT in healing bone injuries that require ultrasonic penetration through a certain distance of tissue. Lee et al. [105] developed a new flexible PMUT array integrated into flexible PDMS. The PMUT arrays were fabricated using a conventional microelectromechanical systems (MEMS) process and bonding techniques for PDMS materials. The flexible PMUT array demonstrated good flexibility and was able to withstand compressive stretching without failure during mechanical bending tests. The PDMS layer provided flexibility in controlling or increasing the gap in the PMUT elements during tensile stretching. The research also explored the prospective utility of ultrasound stimulation utilizing the PMUT array. The findings indicated that the acoustic output pressure achieved sufficient levels for low-intensity ultrasound brain stimulation, registering a sound intensity of 44 mW/cm^2 at 80 V.

2) Promoting Healing: Chronic Wound Management: Lyu et al. [151] have revolutionized the field of ultrasound with their pioneering work on flexible ultrasonic patches for wound healing [see Fig. 5(b)]. By integrating a flexible circuit substrate with piezoelectric ceramics, they have created a lightweight and adaptable patch that conforms seamlessly to treatment areas. This innovative approach, combined with controlled bending and a hydrogel encapsulation layer, holds immense promise for accelerating wound healing through targeted acoustic stimulation. The fabrication process involves meticulous steps, including PMMA deposition, PI application, metal electrode formation, and integration of PZT-4 piezoelectric material. Rigorous acoustic performance evaluation and immunohistochemical experiments validate the patch's effectiveness in activating Rac1 for wound healing. This work represents a transformative advancement, showcasing the potential of flexible ultrasonic patches as a powerful tool in therapeutic ultrasound applications.

3) Healing Hand: Drug Delivery: Addressing the burgeoning demand for enhanced transdermal absorption of therapeutic agents, a pioneering conformable ultrasound patch (cUSP) has been developed by Yu et al. [152] to facilitate controlled drug delivery [see Fig. 5(c)]. Engineered for effective transdermal delivery, this innovative method utilizes piezoelectric transducers integrated within a PDMS substrate. The setup generates intermediate-frequency sonophoresis within a coupling medium positioned between the patch and the skin, facilitating precise therapeutic interventions for different skin concerns and premature aging. Remarkably, the cUSP achieves a remarkable 26.2-fold increase in niacinamide transport in vitro, showcasing its potential for noninvasive treatment applications. Incorporating a strategic arrangement of piezoelectric disks (PZT-Ds) within the PDMS substrate, the cUSP offers controlled spacing and precise alignment for optimal treatment efficiency. The resulting cavity generates localized cavitation pockets, enabling efficient transdermal drug delivery. The novel cUSP design replaces the need for bulky handheld probes, offering patients and consumers a practical and effective method for skin treatment. The integration of meticulous fabrication processes, acoustic spectrum analysis, and thorough characterization underlines the potential of the cUSP as a transformative tool in

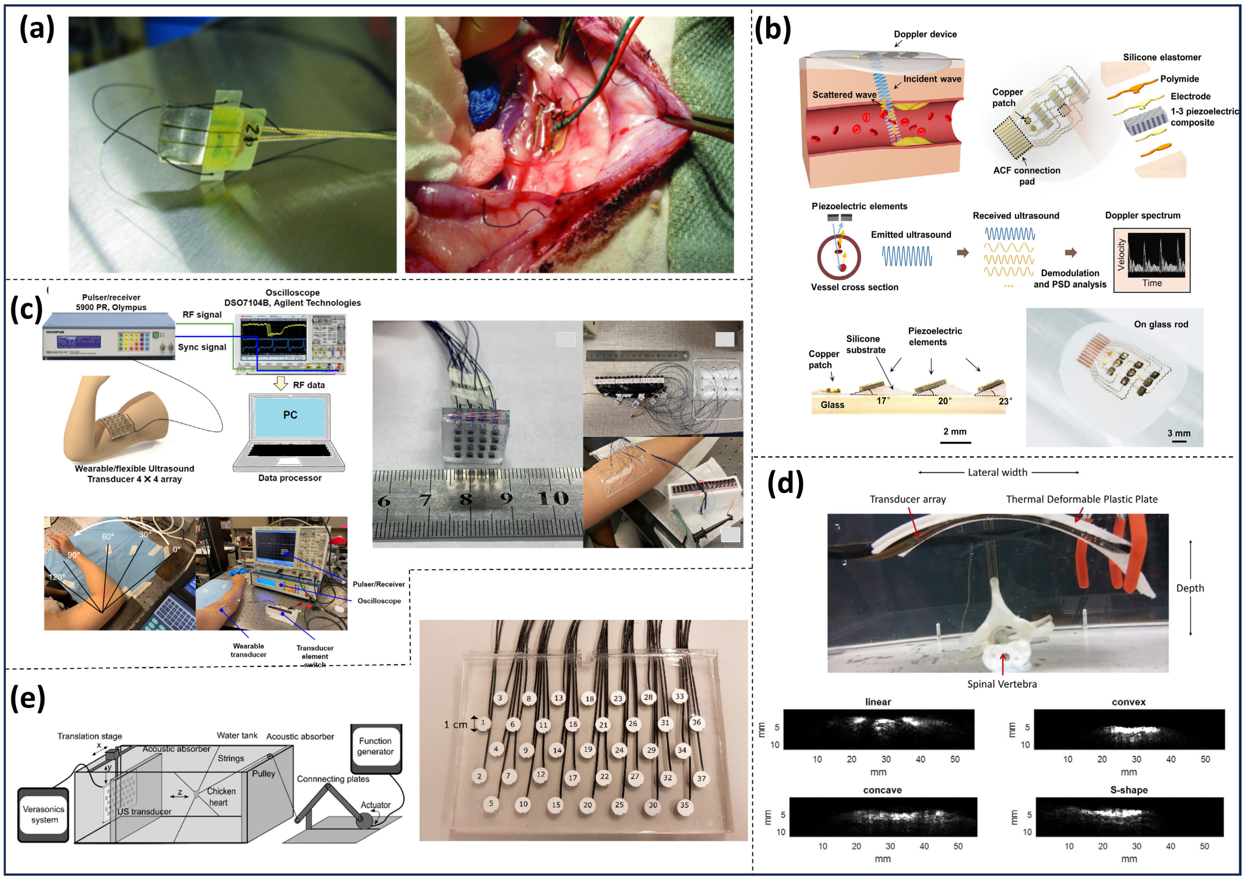

Fig. 4. Flexible transducers for blood vessel measurements and health monitoring. (a) Ultrasound transducer type-flexible piezoelectric polymer with a flexible circuit. A flow sensor prototype that can be placed around a blood vessel (left). The proposed sensor was wrapped around a rabbit infrarenal aorta with a diameter of 2.6 mm (right). Adapted from [141]. (b) Ultrasound transducer type-piezoelectric composite with flexible circuits and substrate. Schematics of the Doppler ultrasonic device and its working principle. Adapted from [142]. (c) Ultrasound transducer type-sparse array with rigid piezoelectric transducers and flexible substrate. Experiment configuration of an in vivo test (left). Photographs of the fabricated flexible transducer and demonstration of its flexibility and wearability (right). Adapted from [145]. (d) Photograph of the experimental configuration used for capturing ultrasound scans from a spine phantom. The flexible transducer is submerged in a water tank, and any bubbles on the transducer surface are meticulously eliminated before imaging (top). The enlarged sections of the images display the chosen region of interest (ROI) of the spinous process, which is produced by the linear, convex, concave, and S-shaped transducer array (bottom). Adapted from [146]. (e) Ultrasound transducer type-sparse array with rigid piezoelectric transducers and flexible substrate. Schematic of the in vitro test setup to monitor the chicken heart rate (left). Photograph of the flexible transducer (right). Adapted from [143].

therapeutic ultrasound and skin health. With a flexible concave lens, the laser-generated-focused ultrasound holds promise for drug delivery owing to its tight focal spot, broad frequency band, and stable excitation with minimal ultrasound-induced heating, which showed a 2.5-fold increase compared with passive release [153].

4) Regulating the Nervous System: Neuromodulation: Pashaei et al. [154] carried out a research endeavor aimed at creating and validating body-conforming ultrasound patches with active capabilities, integrating imaging and modulation modalities for the purpose of image-guided neural therapy. They created a flexible linear array comprising 64 PZT elements, mechanically adaptable, and resonating at 5 MHz to facilitate nerve localization. In addition, a second array of eight elements was designed for low-intensity focused ultrasound neuromodulation at a resonance frequency of 1.3 MHz, utilizing larger PZT elements. The conformal array is based on a flexible PCB layer with signal and ground electrodes on one side. This setup enables personalized feedback on array curvature through a strain sensor on the probe, facilitating real-time adjustments for optimal focusing and image processing. In their subsequent work [155], Pashaei et al. made several improvements, including the use of a custom electrode pattern to simplify assembly and eliminate additional ground layers used in earlier work and solid-state switches for onboard multiplexing to reduce the number of cables and connectors. The individual transducer elements have a spatial resolution of around 1.2 mm, which is sufficient for neural modulation. Fast object-oriented C++ ultrasound simulator (FOCUS) simulations were used to estimate the ability to focus and steer the modulation beam, showing a sensitivity of 80 kPa/V with 3-MHz bandwidth for the modulation. The simulation results suggest that a reasonable transmit voltage of 6 V (i.e., 12 Vpp) is sufficient for generating ∼0.5 MPa required for neuromodulation.

V. Challenges and Future Perspectives

A. Developing Optimal Components for Flexible US Transducers

Creating a flexible ultrasound transducer array tailored for medical imaging applications entails tackling multiple

TABLE V
FLEXIBLE ULTRASONIC TRANSDUCER FOR BIOMEDICAL SENSING

Application	Number of Elements	Center Frequency [MHz]	Thickness [mm]	Size	Weight [g]	Bandwidth [%]	Ref
Blood flow measurement	N/A	40	0.3	12 mm $\times$ 14 mm	$<$ 0.1	$>$ 40	[141]
Blood flow measurement	3$\times$3	5	1.04	1.5 mm$\times$ 1.5 mm per element	0.75	N/A	[142]
Blood pressure monitoring	N/A	5	0.3	6 mm $\times$ 6 mm	N/A	47.6	[20]
Blood pressure monitoring	4 $\times$ 5	7.5	0.24	0.9 mm $\times$ 0.9 mm per element	0.15	32	[17]
Heart rate measurement	37	1	2	1 cm in diameter per element	N/A	N/A	[143]
Breast tumor detection	128	10	N/A	2 mm$\times$ 5 mm per element	N/A	70	[144]
Muscle activity monitoring	4 $\times$4	10	1.2	1.4 mm $\times$ 1.4 mm per element	N/A	61	[98]
Spinal deformity measurement	128	5	N/A	1 mm $\times$ 12 mm per element	N/A	65	[146]

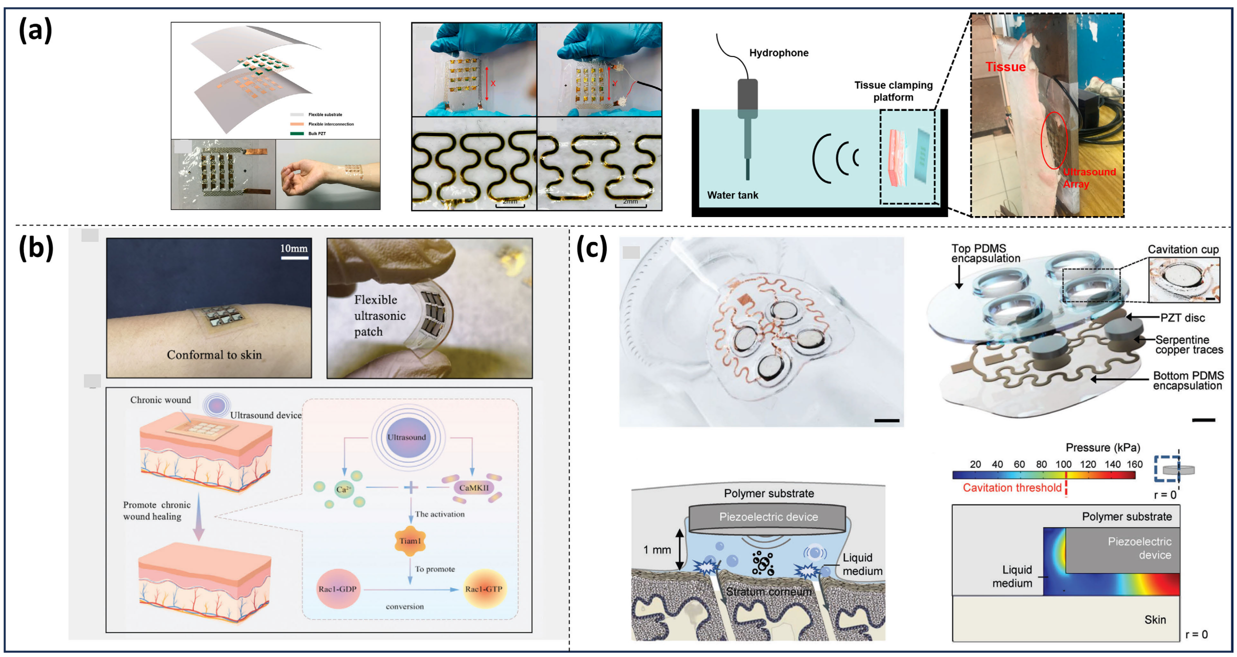

Fig. 5. Flexible transducer for biomedical therapy. (a) Ultrasound transducer type-PMUT arrays. Schematic drawing and photograph of the flexible PMUT (left). The flexibility test of the device (middle). The capability test of ultrasound to penetrate (right). Adapted from [150]. (b) Ultrasound transducer type-rigid piezoelectric materials with flexible interconnect and substrate. Flexible ultrasonic patch design: schematic depiction of the flexible ultrasonic patch's principle, design, and fabrication. The patch seamlessly conforms to skin curvature (bending radius <5 mm) (top). Ultrasound triggers wound healing by activating Rac1 (conversion of Rac1-GDP to Rac1-GTP) (bottom). Adapted from [151]. (c) Ultrasound transducer type-rigid piezoelectric materials with flexible interconnect and substrate. (i) Photograph of a curvilinear glass cylinder with an attached 2-D array of cUSP. (ii) Exploded view of a 2-D array of cUSP. (iii) Schematic showing cUSP on the skin, illustrating cavitation within the device-skin cavity. (iv) Acoustic pressure distribution in radial mode operation. Adapted from [152].

hurdles. Traditional ultrasound transducers are made up of rigid components, including a matching layer, an active layer, a backing layer, and electrodes. To make a transducer flexible, all of these components must be flexible as well. While it is possible to create a flexible US transducer without a backing layer, achieving the desired short-length waveform necessary for high imaging resolution can be difficult. Several attempts have been made to use flexible materials for the backing layer of the transducer array. For example, Kapton PCB has been used in some studies, but its flexibility does not provide sufficient acoustic attenuation properties [155]. In contrast, other studies have used flexible PU rubber with tungsten particles, but this can still hinder the flexibility of the transducer due to the thickness of the backing layer [37]. Finding the optimal material for the backing layer is crucial for the development of a flexible US transducer array that performs well and is also highly flexible. Another challenge is the design of the transducer array itself. The flexibility of the array

TABLE VI
FLEXIBLE ULTRASONIC TRANSDUCER FOR BIOMEDICAL THERAPY

Applications	Number of elements	Center Frequency [kHz]	Size	Acoustic intensity $(\mathrm{mW/cm^2})$	Ref
Bone Injury Treatment	16 elements	321.15	5 mm × 5 mm × 1 mm per element	$I_{\mathrm{sppa}} = 5.5$	[150]
Deep brain stimulation	16 elements	694.4	N/A	$I_{\mathrm{sppa}} = 44$	[105]
Image-Guided Neuromodulation	8 elements	1300	total width 0.94 cm	$I_{\mathrm{sppa}} = 8446$	[154], [155]
Chronic wound Healing	9 elements	2200	4 mm × 4 mm	$I_{\mathrm{sppa}} = 20$	[151]
Drug delivery	4 elements	1300	5 cm × 5 cm × 2 mm	N/A	[152]

transducer is still constrained by the presence of rigid island elements, which hinders its overall flexibility. Therefore, the development of a fully flexible and stretchable US transducer array would be a byproduct of advances in flexible electronics. It is essential to develop optimal flexible components that are suitable for the transducer to improve its design and performance. Future research should focus on overcoming these challenges and developing materials and designs that allow for greater flexibility without sacrificing performance.

While progress has been made in the development of flexible ultrasound transducers, there are still significant challenges to be addressed. Finding optimal materials for the backing layer and designing fully flexible and stretchable transducer arrays are just a few of the challenges that need to be overcome. By addressing these challenges and developing better materials and designs, the potential of flexible ultrasound transducers for biomedical imaging, sensing, and therapy can be fully realized.

B. Acoustic Deformation in the Flexible Transducer Array

The beam profile of the flexible transducer array is crucial for optimal biomedical imaging applications. The beam shape and focusing ability of the transducer are essential for achieving high-quality images with good contrast and resolution. However, due to the nature of flexible transducers, the beam profile can be affected by the curvature of the transducer when it is placed on the target. This can lead to imaging artifacts and reduced image quality.

To address this challenge, researchers have explored novel techniques for beamforming and the integration of ultrasound sensing technology with other modalities. For instance, Huang et al. [156] proposed an end-to-end deep learning approach to solve incorrect time delays, resulting in B-mode imaging with reduced distortion. In addition, Elloian et al. [39] suggested using phase correction algorithms to further enhance image reconstruction on curved surfaces. Chen et al. [133] utilized a laser scanning system to track the position of each element, resulting in a more comprehensive and accurate assessment of the beam profile. Furthermore, the probe incorporates a strain sensor that offers customized curvature feedback for real-time enhancements in focusing and image processing, ultimately leading to enhanced beam profiles [155]. In Chen et al.'s work [157], they integrated an optical shape-sensing (OSS) fiber into a 128-element flexible linear array transducer. The integrated OSS fiber offered dual functionality, enabling real-time tracking of array element positions and deformation. This was achieved by utilizing OSS technology to monitor optical fiber curvature, enhancing the accurate positioning of the array elements. This remarkable capability of OSS technology is underpinned by its proficiency in discerning the curvature of optical fibers through the analysis of acquired backscattered signals, thereby facilitating accurate shape determination of the targeted structure [158]. These novel techniques and integrations are expected to improve the performance and design of flexible US transducers and advance their application.

C. Dry Couplant and Adhesive

The need for a wearable US array is essential for long-term monitoring of the body, but the gel-based coupling medium used in conventional transducers is not suitable for long-term wearability. The use of a gel-typed coupling medium can make the transducer slippery, resulting in unstable experimental RF data for US imaging. Moreover, it is not suitable for long-term wearability because it can irritate the skin and pose hygiene risks. Thus, the adhesive-equipped wearable US transducer with dry couplant emerges as a promising option. Dry couplant offers an alternative to the skin-friendly wearable US transducer, showcasing comparable acoustic impedance to water, minimal acoustic attenuation, and remarkable durability [21]. By using dry couplant and adhesive, a wearable US transducer can be feasible for long-term monitoring of the skin with minimal discomfort. Wang et al. [21] proposed a bioadhesive dry couplant that can provide long-lasting adhesion and can reduce motion artifacts during skin deformation. This technology can overcome the limitations of traditional gel-based coupling media and improve the wearability and user experience of wearable US transducers. Dry couplant and adhesive can be introduced as a promising alternative to the gel-typed coupling medium used in conventional transducers, making the wearable US transducer more comfortable and suitable for long-term monitoring of the skin. The development of skin-friendly dry couplant and adhesive will enhance the user experience and further promote the application of wearable US transducers in various fields.

D. Material Biocompatibility and Safety

Ensuring the biocompatibility and long-term safety of materials employed in flexible ultrasound transducers constitutes a critical imperative for successful clinical translation. The intimate and prolonged interaction between these devices and human tissues mandates the meticulous selection of materials that not only exhibit mechanical flexibility but also showcase attributes of biocompatibility, biofunctionality, and minimal immune reactivity. Diverse materials, ranging from polymers and hydrogels to elastomers, have exhibited promise in the realm of flexible transducer design. However, their appropriateness for sustained implantation or prolonged usage necessitates a comprehensive evaluation.

The pursuit of biocompatible materials has instigated exploration into innovative avenues, encompassing bioactive coatings, bioresorbable polymers, and tissue-engineered constructs. Notably, biodegradable polymers, such as polylactic acid (PLA) and poly(lactic-co-glycolic acid) (PLGA), have garnered attention due to their controlled degradation kinetics, offering the potential for mitigating long-term inflammation [159]. Furthermore, ongoing research underscores the realm of flexible biodegradable piezoelectric polymer materials. This progress has led to the development of efficient piezoelectric PLLA nanofibers, an example of biocompatible and sustainable piezoelectricity [35]. Another innovative avenue embraces the incorporation of amino acid crystals, specifically glycine, within PCL nanofibers. This approach yields flexible, biodegradable, and piezoelectric glycine-PCL nanofiber films, surpassing existing biodegradable transducers in terms of piezoelectric efficiency and biocompatibility [51].

Anticipating the future, the integration of smart materials, responsive to physiological cues, such as pH, temperature, or enzymatic activity, holds potential for elevating biocompatibility and safety through dynamic interactions with the host environment [160]. By effectively addressing the challenges surrounding biocompatibility and safety, the evolution of flexible ultrasound transducers is poised to harmoniously assimilate into biological systems, thereby enhancing clinical viability and ultimately fostering improved patient outcomes.

E. Wearable Integration, Patient Comfort, and Real-Time Monitoring

The seamless integration of flexible ultrasound transducers into wearable devices poses a formidable challenge while intersecting with the imperatives of patient comfort and real-time monitoring. The optimization of wearable ultrasound systems mandates a delicate equilibrium between acquiring accurate data and ensuring the wearer's ease. Striking this equilibrium necessitates not only the engineering of devices for conformability and stability but also the integration of advanced soft materials and innovative attachment methodologies. A design approach rooted in ergonomic principles and centered on the user experience is pivotal to guaranteeing prolonged wearer comfort, ultimately enhancing compliance and enabling continuous monitoring. The vitality of real-time monitoring and precise data interpretation cannot be overstated within the realm of wearable ultrasound systems. Overcoming hurdles in data processing, artifact mitigation, and on-device computation is indispensable for the seamless assimilation of flexible ultrasound devices into point-of-care diagnostics and continuous health monitoring ecosystems. The integration of advanced signal processing algorithms and cutting-edge machine learning techniques assumes a central role in the extraction of meaningful insights from acquired ultrasound data, thus facilitating prompt and well-informed clinical decisions.

An illustrative example that embodies the convergence of these challenges and future directions is the work presented in Lin et al. [126]. This groundbreaking study introduces a wearable USoP, featuring a compact and flexible control circuit that interfaces with an ultrasound transducer array. Machine learning algorithms are utilized for the purpose of monitoring moving tissue targets and enhancing data analysis. The capabilities of the USoP include ongoing monitoring of physiological signals from tissues located up to a depth of 164 mm. This enables the surveillance of central blood pressure, heart rate, and cardiac output in mobile subjects over extended periods. This exemplar not only underscores the significance of wearable integration and real-time monitoring but also showcases the transformative potential of addressing these challenges through interdisciplinary innovation.

The overarching theme within this challenging segment revolves around the development of wearable ultrasound systems that not only seamlessly meld with human physiology but also provide meaningful and real-time data insights, making them indispensable tools in the realm of modern healthcare.

VI. Conclusion

The purpose of this narrative review article was to summarize recent technological developments in flexible US transducers. We have explored advanced materials and fabrication techniques for flexible US transducers, along with recent applications of these transducers in the fields of biomedical imaging, sensing, and therapy. Research and development challenges and future perspectives are discussed. It is clear that flexible US transducers are a promising technology for biomedical imaging, sensing, and therapeutical applications. They offer the capability to deliver enhanced performance in contrast to conventional rigid transducers and are applicable across a diverse spectrum of biomedical uses. However, there are still many technical barriers and challenges to overcome in order to fully translate these potential devices. With further investigations and developments, flexible US transducers will surely become a critical tool in the biomedical field, improving our understanding of human pathophysiology and the treatment of various diseases.

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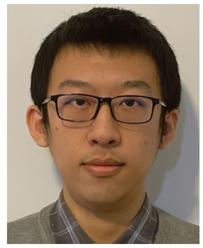

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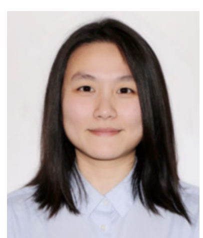

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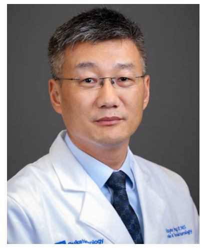

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