Fetal monitoring for high-risk pregnancies using a wearable ultrasound patch

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Ultrasonography is a cornerstone of fetal monitoring, but it requires specialized sonographers and is limited to providing snapshot evaluations at discrete intervals. Here we report a wearable ultrasound patch (UPatch) for continuous and autonomous fetal monitoring. The UPatch can acquire anatomical structures and blood flow velocities, demonstrating high agreement with a handheld clinical ultrasound device on 62 pregnancies. Real-time image segmentation allows autonomous tracking of target vessels and thus the acquisition of continuous blood flow spectra during fetal and maternal movements without a sonographer. Continuous monitoring data for 1–3 h from 52 pregnant women aligned with stratified perinatal conditions, including healthy, small for gestational age, large for gestational age, gestational diabetes, pre-eclampsia, and gestational hypertension. Additionally, integration with a miniaturized circuit could enable fully wireless operation and greater patient mobility. The UPatch could provide evidence of fetal complications in high-risk pregnancies, expanding prenatal-care capabilities.

The physiological maturation of the fetus in the uterus is supported by cardiovascular exchanges between the maternal, placental, and fetal circulations[1]. Any disruption of this delicate exchange system can precipitate fetal complications, such as hypoxia[2], intrauterine growth restriction[3], and cerebral palsy[4], with stillbirth[5] being the most extreme outcome. Despite decades of research, many such fetal complications remain undiagnosed and unexplained, with existing methods relying mostly on postmortem investigations[6,7]. Continuous monitoring of fetal anatomy and hemodynamics throughout pregnancy not only provides insights into the pathophysiological mechanisms of fetal complications but also allows the early detection and

mitigation of emerging issues, helping to prevent adverse outcomes (Supplementary Discussion 1)[8,9].

Continuous fetal monitoring currently relies predominantly on cardiotocography, which measures the fetal heart rate and uterine contractions (Supplementary Fig. 1 and Supplementary Discussion 2)[10,11]. However, given the complexity of fetal physiology, these measurements do not correlate well with fetal health and yield false positive rates up to 50% (Supplementary Discussion 3)[12-15]. Other continuous monitoring devices based on electrocardiography[16], pulse oximetry[17], or accelerometry[18] cannot reliably differentiate fetal from maternal signals and are not commonly used in clinical practice. Ultrasonography is widely used to measure fetal biometry and blood flow (Supplementary Discussion 4)[19-21], and standardized reference indices have been established throughout gestation to evaluate fetal health (Supplementary Discussion 5)[22-25]. However, conventional ultrasonography is heavily dependent on the skills of the operator, and measurements are limited to snapshots that can fail to capture the patterns and critical moments of fetal hemodynamics[26,27]. Although wearable ultrasound devices have been developed for continuous monitoring of physiological signals[28-31], they are unsuitable for fetal monitoring due to weak blood flow signals from deep fetal vessels and signal loss during large fetal movements.

Here we report a wearable ultrasound patch (UPatch) that can continuously image the fetus and autonomously measure fetal blood flow in real time (Supplementary Discussion 6). To enhance the signal-to-noise ratio and imaging resolution, we integrated transducers fabricated by super multipass dicing, as well as an acoustic lens and a soft Faraday cage (Extended Data Fig. 1). To identify and track moving targets, we used segmentation-based algorithms to place sample gates in vessel images. Blood flow spectra can be acquired continuously and autonomously in

real time, even during large fetal movements. We conducted a prospective study with 62 pregnant participants to validate the accuracy of the UPatch using a handheld clinical ultrasound device. Furthermore, we used the UPatch to continuously monitor the umbilical cord blood flow in 52 participants, generating longitudinal measurements that distinguished transient fluctuations from sustained compromise. We found that Doppler indices exhibited monotonic relationship with gestational age and differentiated high-risk from healthy pregnancies. In a pre-eclamptic case, the UPatch revealed the severity of intrauterine growth restriction, leading to Cesarian delivery to prevent stillbirth.

Design of the UPatch

Fetal structures and vessels vary in size and are distributed over a wide range of depths in the uterus (Fig. 1a, left)[32]. Brightness-mode imaging is based on echoes from tissue interfaces and distributed scattering within tissues[33]. On the other hand, color Doppler imaging and spectral Doppler waveforms of blood flow are based on the much weaker echoes from red blood cells[33,34].

We adopted three strategies to strengthen the echoes from red blood cells in deep fetal vessels. First, the UPatch contains a 2.5-MHz PZT-5H 1–3 composite with a conductive epoxy backing layer at a pitch of 0.65 ultrasound wavelength (Fig. 1a, right). We used super multipass dicing, which reduced the kerf and thus increased the element width by 58% (Fig. 1b, Methods, Supplementary Figs 2 and 3, and Supplementary Discussion 7). This improved the signal-to-noise ratio by 7.2 dB without increasing the crosstalk between adjacent elements (Supplementary Fig. 4). Second, we designed an acoustic lens to improve elevational focusing, which reduced the focal beamwidth by 25% and thus enhanced the spatial resolution and color Doppler signal

sensitivity (Fig. 1c, Methods, Supplementary Figs 5–8, and Supplementary Discussion 8)[35]. Additionally, the acoustic lens moved the beam elevational focal depth from 14 to 10 cm, on par with clinical devices, to balance the spatial resolution throughout the entire uterus (Fig. 1c and Supplementary Fig. 7). Third, we developed a soft Faraday cage by connecting the electromagnetic shielding layer to the mesh ground electrode layer using vertical interconnect accesses (Methods, Supplementary Figs 9 and 10, and Supplementary Discussion 9). The cage fully encapsulated the elements and reduced the electromagnetic interference by 11.7 dB/Hz (Fig. 1d and Supplementary Figs 11–13)[36]. Additionally, we used two signal electrode layers that can individually address each element with a minimal electrode footprint (Supplementary Fig. 9). The UPatch was mechanically compliant (Fig. 1e) and showed negligible constraints on the movement of pregnant participants in various positions (Supplementary Fig. 14).

To quantify the UPatch performance, we characterized its brightness, color Doppler, and spectral Doppler modes following the Institute of Physics and Engineering in Medicine protocols (Methods)[37]. On a phantom filled with monofilament wires and tissue-mimicking contrast targets, the axial and lateral resolutions, dynamic range, and contrast-to-noise ratio of the UPatch in brightness mode were comparable to those of a handheld clinical ultrasound device (Supplementary Fig. 15)[38]. On a flow phantom with blood-mimicking fluids, the color Doppler sensitivity and the spectral Doppler accuracy of the UPatch at different depths and velocities were similar to those of the clinical device (Supplementary Figs 16 and 17)[39]. On a customized umbilical cord flow phantom, the UPatch reliably acquired flow signals regardless of phantom orientation (Extended Data Fig. 2, Supplementary Figs 18 and 19, and Supplementary Discussion 10). On a healthy adult participant, all three modes of the UPatch and the clinical device were comparable in quality (Supplementary Figs 20 and 21).

125

126 **Acoustic safety characterizations**

127 Minimal acoustic intensity is necessary for the fetus to avoid potential side effects
128 associated with ultrasound exposure[40]. We examined the biological effects of the UPatch in vitro
129 in all three modes following the guidelines from the American Institute of Ultrasound in
130 Medicine (AIUM)[41], British Medical Ultrasound Society (BMUS)[42], and Food and Drug
131 Administration (FDA)[43] (Methods and Supplementary Discussion 11). The maximum spatial
132 average temporal average intensity was 0.72 mW/cm^2 (< 20 mW/cm^2 is recommended by the
133 FDA[43], and 5 mW/cm^2 is the value reported for the GE Corometrics 250CX[44]). Furthermore, the
134 maximum derated spatial peak pulse average intensity was 16.42 W/cm^2 (< 190 W/cm^2 is
135 recommended by the FDA[43]), the maximum derated spatial peak temporal average intensity was
136 58.5 mW/cm^2 (< 94 mW/cm^2 is recommended by the FDA[43]), and the mechanical index was 0.20
137 for duplex (combined brightness and color Doppler) imaging and 0.38 for spectral Doppler (<
138 1.9 is recommended by the FDA[43] and ≤ 0.7 is recommended by the BMUS[42]) (Fig. 1f and
139 Supplementary Discussion 12).

140 We also evaluated the thermal characteristics of the UPatch[45,46]. Surface temperatures
141 were tested using a thermocouple on a phantom and in air (Methods). In both experiments, the
142 temperature increase was < 0.7 °C during 48 h of continuous UPatch activation (Supplementary
143 Fig. 22). To calculate the thermal index, acoustic power was measured and cross-checked using a
144 hydrophone and radiation force balance (Methods, Supplementary Figs 23 and 24)[46]. The soft
145 tissue thermal index, bone thermal index, and cranium thermal index were 0.02, 0.05, and 0.05
146 for duplex imaging, and 0.1, 0.38, and 0.17 for spectral Doppler, respectively (Fig. 1f). All

results were well below the maximum level of 0.7 recommended by the AIUM and BMUS for continuous monitoring[41,42].

Fetal monitoring and validation

Duplex imaging can reveal fetal anatomical abnormalities, quantify fetal biometry, and facilitate the recording of spectral Doppler waveforms[47]. Correcting the phase aberration induced by the maternal abdomen allowed the acquisition of high-quality signals from the UPatch (Supplementary Fig. 25 and Supplementary Discussion 13). The UPatch was able to image the two arteries and a vein in the helical umbilical cord and the middle cerebral artery from the circle of Willis (Fig. 2a,b). Detailed blood flow velocities were acquired from all of these vessels (Fig. 2c and Supplementary Discussion 5). The cerebroplacental ratio (the pulsatility index of the middle cerebral artery divided by that of the umbilical artery) quantifies the fetal brain-sparing effect during prolonged hypoxia or in placental insufficiency, and the measurements acquired by the UPatch and a clinical device were comparable (Fig. 2d)[48]. The UPatch accurately measured anatomical structures (such as the biparietal diameter, head circumference, abdominal circumference, and femur length), providing a fetal weight estimate for the diagnosis of growth-related complications (Fig. 2e, Methods, and Supplementary Fig. 26)[49].

The UPatch accuracy was validated against a clinical device on 62 participants with various vessel depths, orientations, and placental locations (Table 1, Methods, Extended Data Fig. 3, and Supplementary Figs 27 and 28). From the spectral Doppler waveforms, an envelope extraction algorithm can derive peak systolic velocity, end diastolic velocity, and fetal heart rate to calculate the systolic-to-diastolic ratio, pulsatility index, and resistance index (Supplementary Fig. 29)[50]. Based on Bland–Altman analysis, the mean differences ± standard deviations between

the UPatch and clinical device were –0.019 ± 0.274 for the systolic-to-diastolic ratio (–0.72% relative mean difference), –0.010 ± 0.117 for the pulsatility index (–0.92%), –0.004 ± 0.044 for the resistance index (–0.64%), and 0.95 ± 2.91 beats per minute (bpm) for the fetal heart rate (0.68%; Fig. 2f and Supplementary Fig. 30). Pearson correlation coefficients demonstrated strong linear relationship between the two devices (Extended Data Fig. 4).

Autonomous vessel tracking

The umbilical cord facilitates gas exchange and nutrient transfer between the placenta and fetus[1]. Spectral Doppler waveforms of the umbilical blood flow can monitor the fetoplacental circulation (Supplementary Discussions 4 and 5)[3]. However, the umbilical cord is a moving target due to fetal and maternal movements, respiration, and uterine contractions (Supplementary Discussion 14)[21]. In clinical practice, sonographers manually adjust the color-flow box and set a sample gate to acquire spectral Doppler signals from the target vessel, which is time-consuming and operator-dependent (Supplementary Fig. 31)[51]. In low-resource settings where the incidence of pregnancy complications remains high, such sonographers are often limited in number[52,53].

To reduce the need for manual operators and the corresponding healthcare burden, we developed an image segmentation-based algorithm that can autonomously identify and track moving vessels (Fig. 3a and Supplementary Discussion 15). We used a diverging beam to cover a large field of view, ensuring that the moving vessel remains consistently visible within the ultrasonographic window (Supplementary Fig. 32)[54]. The UPatch stably acquired the umbilical cord signals without repositioning across common maternal postures (Extended Data Fig. 5). By segmenting the reconstructed duplex images, the tracking algorithm identifies pulsatile signals

by calculating the variance in color Doppler pixel intensity over several consecutive frames (Fig. 3a, Extended Data Fig. 6, and Supplementary Discussion 16). Then, the algorithm identifies the primary region of the pulsatile signals and registers the spatial centroid of the primary region as the sample gate, where a focused beam is directed to acquire blood flow spectra with enhanced signal intensity (Fig. 3a and Supplementary Figs 33 and 34)[54]. This autonomous sample gate registration enabled continuous blood flow monitoring of a moving vessel in real time (Fig. 3b,c, Supplementary Video 1, and Supplementary Discussion 17). Without the tracking algorithm, blood flow signals were lost during vessel movements (Fig. 3c).

To validate the accuracy of the autonomous tracking algorithm, we compared its sample gates with those registered by a sonographer in a double-blind test (Fig. 3d, Methods, and Supplementary Discussion 18). We found that 91.9% and 90.5% of the sample gates were within a 2-mm discrepancy (an accepted clinical standard) in the lateral and axial directions, respectively (Fig. 3e and Supplementary Discussion 19)[47]. In a separate test, the sample gates registered by the tracking algorithm were considered accurate by three sonographers in 94.2% of the images acquired using the clinical device (with some images affected by fast cord movements), and in 91.2% of the images acquired using the UPatch (due to slight differences in image quality, Methods, Extended Data Fig. 7, and Supplementary Discussion 18). We used Bland–Altman analysis to compare the systolic-to-diastolic ratio from the sample gates acquired by the tracking algorithm and a sonographer (Methods and Supplementary Discussion 18). The mean difference ± standard deviation was only 0.036 ± 0.124 (Fig. 3f). While its accuracy is similar to that of U-Net[55] and ConvLSTM[56], the tracking algorithm achieved more than a tenfold reduction in processing time, enabling real-time image processing (Supplementary Fig. 35 and Supplementary Discussion 18).

Continuous monitoring of pregnant participants

The continuous monitoring of umbilical artery blood flow is important for assessing fetal growth restriction (Supplementary Discussions 4 and 5)[57]. Abnormal waveforms (such as absent or reversed end diastolic velocities) often indicate fetal pathology due to placental insufficiency, and even subtler abnormalities (such as increased Doppler indices) indicate an elevated risk[58,59]. Continuous monitoring of these changes can inform the optimal delivery time, helping avoid both premature birth, which may lead to postnatal complications[60], and delayed birth, which heightens the risk of asphyxia, cerebral palsy, and stillbirth[61].

We used the UPatch to compare a healthy (32 weeks 6 days) and a pre-eclamptic (28 weeks 3 days gestational age) participant. The participants were in a semi-recumbent position with the UPatch placed on the maternal abdomen to monitor the blood flow continuously at the placental cord insertion (Fig. 4a and Supplementary Discussion 20). We derived the fetal heart rate and Doppler indices, including resistance index, pulsatility index, and systolic-to-diastolic ratio, which are independent of the ultrasound incident angle (Supplementary Discussion 21)[19]. Both participants showed similar fetal heart rate patterns (Fig. 4b–d). The average fetal heart rates were 149.5 bpm for the healthy and 143.3 bpm for the pre-eclamptic participant, with variabilities of 20.3 and 21.5 bpm, respectively (Fig. 4b and Supplementary Discussion 22). The UPatch could detect transient fetal heart rate accelerations, which indicate a healthy fetal state (Fig. 4c and Supplementary Fig. 36)[62]. However, the Doppler indices of the participants showed distinct features. There was minimal variability in the Doppler indices of the healthy participant, indicating stable blood flow. The mean resistance index of 0.62, pulsatility index of 1.04, and systolic-to-diastolic ratio of 2.61 aligned with the 50th percentile of established reference norms,

indicating a healthy fetus[23]. In contrast, the pre-eclamptic participant showed large fluctuations in the Doppler indices, and the end diastolic velocity was absent 24.9% of the time (Fig. 4b,d, Supplementary Fig. 37, and Supplementary Discussion 22). The mean resistance index of 0.85, pulsatility index of 2.04, and systolic-to-diastolic ratio of 5.11 were above the 97.5th percentile, suggesting severe placental dysfunction[23]. Following the detection of compromised fetal health using the UPatch, the pre-eclamptic participant underwent intensive monitoring, and the baby was delivered via Cesarean section 4 days later.

Current obstetric practice relies on a two-step process of cardiotocography for fetal heart rate and separate manual Doppler ultrasound for Doppler indices (Supplementary Discussion 23)[57]. We used the UPatch to continuously monitor both signals simultaneously in 52 pregnancies (Supplementary Figs 36–91 and Supplementary Discussion 24). Cohort analysis showed that the fetal heart rate and Doppler indices are not correlated, highlighting their potential to provide complementary clinical insights (Fig. 4e and Supplementary Fig. 92). Additionally, continuous monitoring results revealed that transient physiological fluctuations can temporarily alter Doppler indices to abnormal values (Supplementary Figs 36–91). With conventional ultrasound devices, these transient fluctuations may be misinterpreted as pathological and trigger unnecessary interventions, whereas continuous monitoring with the UPatch establishes a personalized baseline and temporal context to distinguish transient fluctuations from sustained compromise. While fetal heart rate decelerations were observed, the substantial beat-to-beat variability and absence of a defined gestational reference trend make interpretation challenging (Fig. 4f and Supplementary Fig. 93)[63]. In contrast, Doppler indices exhibited monotonic relationships with gestational age, consistent with established reference ranges[23], supporting their use as objective, clinically actionable information for assessing fetal vascular resistance and identifying high-risk pregnancies

(Fig. 4g and Supplementary Figs 93 and 94). The data were also stratified by perinatal conditions, including healthy, small for gestational age, large for gestational age, gestational diabetes, pre-eclampsia, and gestational hypertension. Fetal heart rate distributions overlapped substantially among all conditions, showing minimal discrimination (Fig. 4h). In contrast, Doppler indices differed, particularly for small for gestational age, pre-eclampsia, and hypertension, illustrating the UPatch's potential to identify high-risk pregnancies in real time (Fig. 4i, Supplementary Figs 95 and 96, and Supplementary Discussion 25).

Discussion

The UPatch can visualize fetal anatomy and measure fetal heart rate and blood flow with high sensitivity and accuracy, offering complementary insights to existing approaches to evaluate fetal health. Whereas standard clinical regimens mandate periodic blood flow assessments performed by sonographers, typically at weekly intervals in outpatient settings and up to three times per week for inpatients, these evaluations are labor-intensive and prone to errors[58]. In contrast, the UPatch adheres directly to the maternal abdomen for hands-free use and employs an image segmentation algorithm to differentiate and track blood vessels autonomously, enabling continuous and accurate blood flow measurements without the need for a sonographer.

Several strategies could be pursued to improve the UPatch further. First, the tracking algorithm cannot generate uninterrupted spectral Doppler waveforms during motion because vessel tracking and spectral Doppler acquisition occur sequentially. This problem can be solved by integrating a separate image segmentation algorithm to detect signal loss in real time, allowing the system to pause the spectral Doppler sequence and track the target vessel in the duplex image. Leveraging advanced graphics processing units would substantially accelerate

computation, enabling simultaneous duplex imaging and spectral Doppler acquisition (Supplementary Discussion 17)[64,65].

Second, the UPatch currently needs to be wired to a backend system (Vantage 256, Verasonics) for both power and high-bandwidth data transmission, like many existing wearable devices[28,31,66-70]. While this configuration is appropriate for inpatient settings, it limits mobility and broader applicability. Future developments could incorporate wireless circuits to expand the range of clinical scenarios. A compact circuit was recently developed to interface a transducer array with a single transceiver for vessel pulsation monitoring, but it is not capable of Doppler imaging (Supplementary Discussion 26)[71]. Developing a more advanced circuit with multiple transceivers for Doppler imaging would enhance patient mobility, help assess the impact of maternal posture and physical activity on fetal health, and show how maternal circulation affects fetal hemodynamics[72].

Third, further investigations are necessary to establish the UPatch as a diagnostic tool to identify specific conditions or diseases[73]. Continuous hemodynamic monitoring with a particular focus on the frequency and duration of critical changes (such as decreased or absent end diastolic velocity), is essential for assessing the severity of hypoxia and identifying optimal intervention windows to prevent adverse perinatal outcomes[74]. The UPatch can define individualized baselines that characterize the normal range of short-term fluctuations for each patient while highlighting patterns of sustained abnormalities that indicate evolving pathology. This personalized calibration reduces false alarms triggered by normal transient events and allows clinicians to focus on persistent deviations that truly indicate risk. These studies could establish the real-world clinical utility of the UPatch.

Finally, this study was limited to using only the UPatch. Future work could focus on integrating UPatch data streams (such as fetal heart rate and Doppler indices) with additional physiological metrics from both the fetus (such as electrocardiogram) and the mother (such as blood pressure, pulse oximetry, and tocodynamometry). Such multimodal integration would provide a more comprehensive and accurate understanding of the dynamic interactions between fetus and mother[11]. Additionally, applying big data analysis to correlate UPatch data streams with postnatal and postmortem results would yield unprecedented insights into the mechanisms of fetal complications, advancing diagnosis and intervention strategies[75].

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Author contributions

G.P., A.G., and S.X. conceived the project. G.P., Y.B., H.H., S.Y.Q., M.Y.L., S.W., P.B., G.G., R.T.W., A.X.R., C.C.F.L., X.X.G., H.J.H., A.Y., M.L., and S.X. developed the UPatch. G.P., X.Y.Y., X.J.C., W.T.Y., R.S.W., and K.A.W. carried out the acoustic safety characterization. G.P., Y.B., H.H., A.R., M.T., J.L., A.T.P., L.I., S.L.C., L.C.L., and A.G. performed the clinical studies. G.P., Y.B., H.H., S.Z., R.S. and S.X. processed and analyzed the data. G.P. H.H., A.L., L.S., and M.S. prepared the figures. G.P. and S.X. wrote the paper. All authors provided constructive and valuable feedback on the manuscript.

Competing interests

S.X. is a cofounder of Softsonics LLC. M.T., L.I., and A.G. are cofounders of Safer Birth LTD. A.T.P. is a Senior Scientific Advisor of Intelligent Ultrasound Ltd. All other authors declare no competing interests.

Table 1 | Participant demographics in the validation study. The demographic information includes race/ethnicity, age, height, weight, body mass index, gestational age, and placental location at the time of the study. The validation was performed on a diverse participant cohort, ensuring the broad applicability of the UPatch.

Race/ethnicity	**n (percentage)**
White	45 (72.6%)
Asian	5 (8.1%)
Mixed, American Indian	8 (12.9%)
Unknown*	4 (6.4%)
At the time of study	**Mean ± standard deviation**
Age (years)	31.45 ± 4.88
Height (cm)	164.44 ± 7.34
Weight (kg)	73.33 ± 18.11
Body mass index	27.17 ± 6.92
Gestational age (days)	238.90 ± 26.68
Placental location	**n (percentage)**
Anterior	36 (58.1%)
Posterior	17 (27.4%)
Lateral	4 (6.5%)
Fundal	5 (8.1%)

*The participants refused to provide this information.

Figures and figure legends

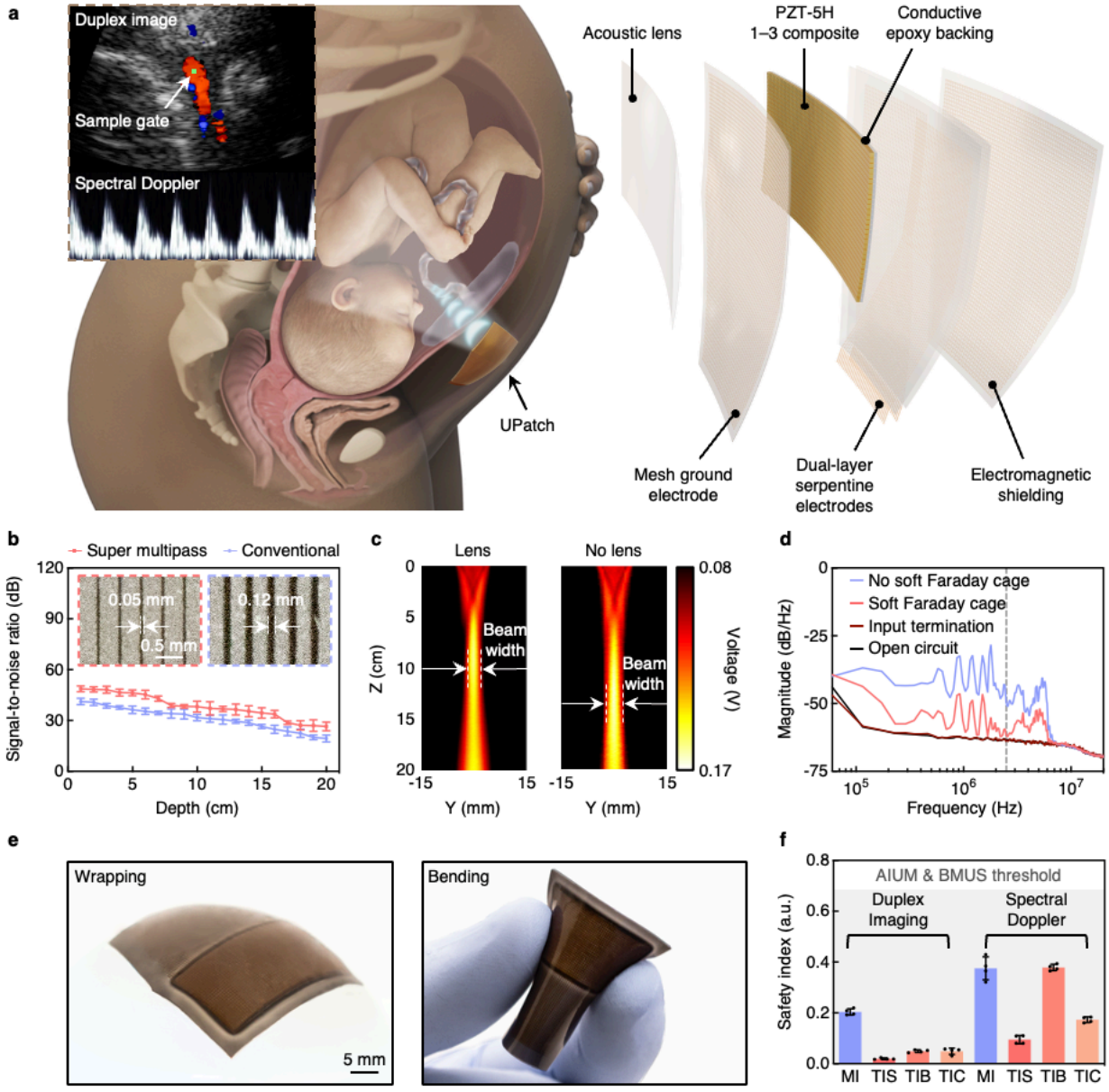


Fig. 1 | Overview of the UPatch. a, Schematics of the working principle for fetal monitoring (left) and a structurally exploded view (right). The UPatch is attached to the maternal abdomen to monitor fetal anatomical structures and hemodynamics continuously with duplex imaging. A sample gate, indicated by the green dot, provides blood flow measurements from fetal vessels. A diverging beam is used for imaging, and a focused beam for blood flow

measurement. The UPatch is multilayered to enhance its signals from red blood cells and minimize its form factors. **b**, Comparison of signal-to-noise ratios between the conventional method and transducers diced with the super multipass method (n = 12). The insets are optical images showing the results from the two methods. The images share the same scale bar. **c**, Acoustic fields measured using a hydrophone in the elevational plane of the UPatch with (left) and without (right) an acoustic lens. **d**, The power spectral density analysis of the UPatch with and without the soft Faraday cage, input termination at 50 Ω, and open circuit of the backend system. The soft Faraday cage reduces noise coupling from electromagnetic interference by 11.7 dB/Hz at 2.5 MHz (gray dashed line), the center frequency of the UPatch. **e**, Photographs of the UPatch demonstrating its mechanical compliance in wrapping (left) and bending (right) shapes. **f**, UPatch safety measurements. The mechanical index (MI), soft tissue thermal index (TIS), bone thermal index (TIB), and cranium thermal index (TIC) were all well below the thresholds of 0.7 recommended by the AIUM[41] and BMUS[42] for continuous monitoring (n = 4). Data in **b** and **f** are presented as mean ± standard deviations.

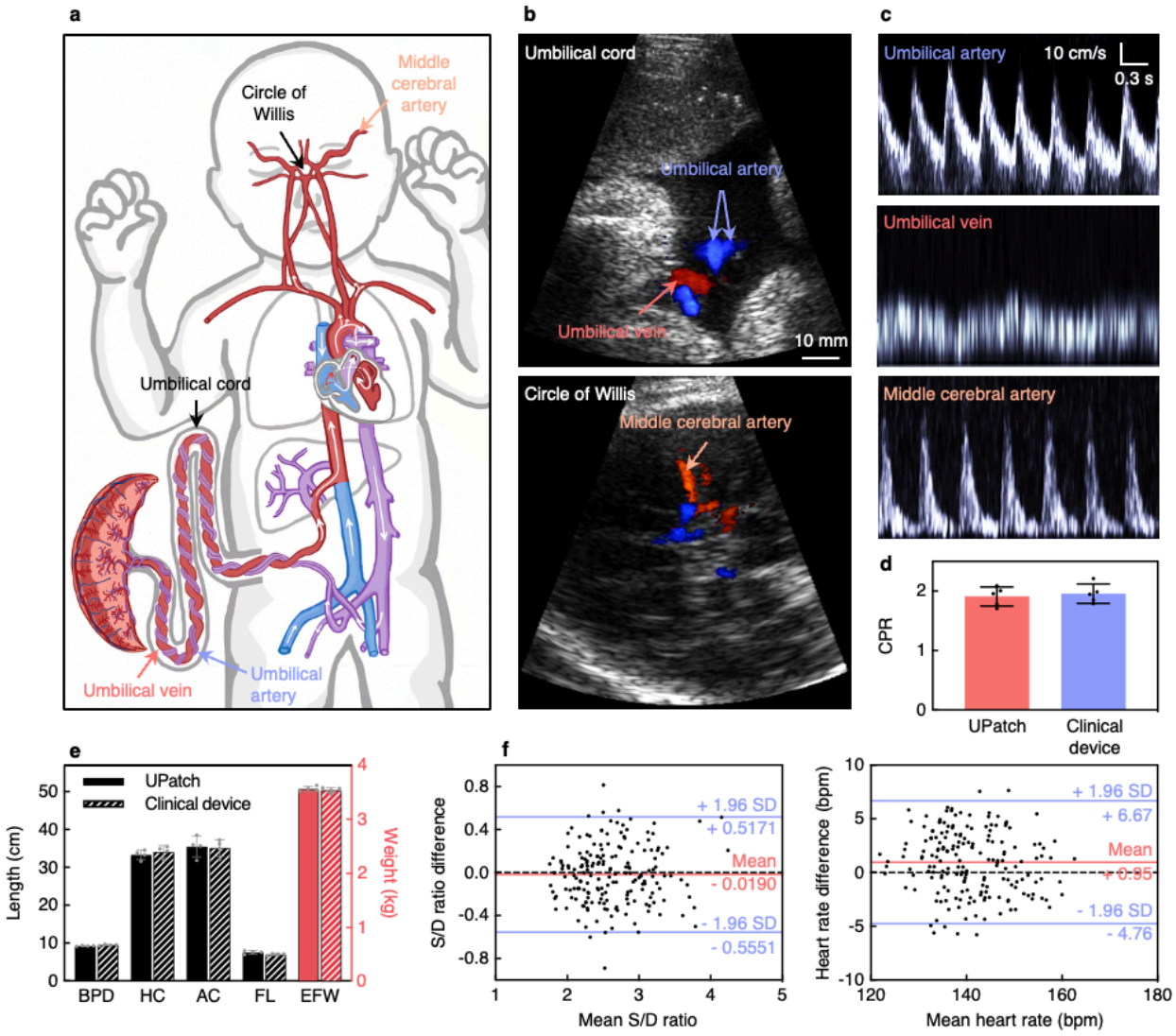
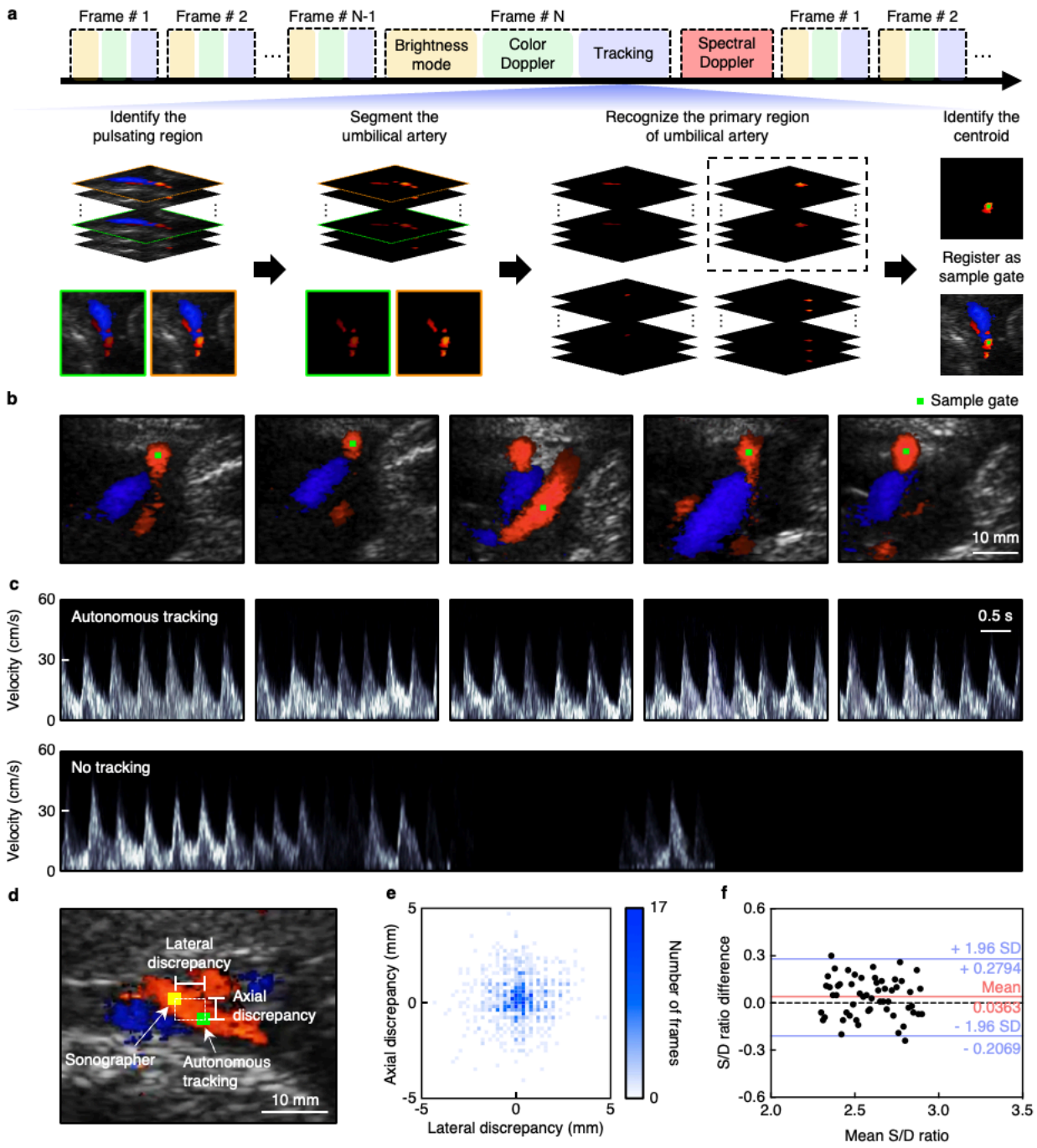


Fig. 2 | Measurements and validations of the UPatch for fetal monitoring. a, Schematics illustrating the fetal cardiovascular system. The umbilical cord shows coiling of the two umbilical arteries and the umbilical vein. The middle cerebral artery is located within the circle of Willis. **b**, Duplex images from the umbilical cord and circle of Willis recorded with the UPatch. The images share the same scale bar. **c**, Blood flow spectra from three key vessels recorded with the UPatch. The spectra share the same scale bar. **d**, Comparison of the cerebroplacental ratio (CPR) measured with the UPatch and a handheld clinical ultrasound device (n = 5). **e**, Comparison of fetal biometry measured with the UPatch and a handheld

clinical ultrasound device. Biparietal diameter (BPD), head circumference (HC), abdominal circumference (AC), and femur length (FL) were used to calculate estimated fetal weight (EFW) (n = 4)[49]. **f**, Bland–Altman plots for the systolic-to-diastolic ratio (S/D ratio, left) and fetal heart rate (right) measured using the UPatch and a handheld clinical ultrasound device (Voluson E10, GE). Solid red lines are the mean differences between the two devices, solid blue lines are 95% limits of agreement (1.96 standard deviations above and below the mean differences), and black dashed lines are the zero difference between the two devices. Each plot has three measurement pairs by repeating three times with the same devices on the same participant for each of the 62 participants. Data in **d** and **e** are presented as mean ± standard deviations.



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394 **Fig. 3 | Autonomous vessel tracking by image segmentation. a**, Beamforming

395 sequence and working principle of the image segmentation-based vessel tracking algorithm. The

396 tracking algorithm is integrated between the duplex imaging and spectral Doppler (top). The

397 algorithm identifies the pulsating regions in consecutive frames, segments the umbilical artery,

recognizes the primary region, and registers the spatial centroid as the sample gate (bottom). Images with green and orange boundaries represent the end diastole and peak systole, respectively. In these two images, the blue regions are similar, indicating the umbilical vein, whereas the red regions differ in intensity, indicating the pulsating umbilical artery. Among the four segmented areas, the primary region is defined as the largest segmented area (denoted with the black dashed box), which has the strongest signal intensity. **b**, Tracking a moving umbilical artery and registering a sample gate using the UPatch with the autonomous algorithm. All images share the same scale bar. **c**, The autonomous algorithm allows the measurement of blood flow from a moving vessel continuously (top). Without the autonomous tracking algorithm, a predefined sample gate results in signal loss due to vessel movements (bottom). The spectra share the same scale bar. **d**, A representative image showing the difference between the sample gate labeled by the autonomous tracking algorithm and by a sonographer. **e**, Summary of the lateral and axial discrepancies in the sample gates labeled by the autonomous tracking algorithm and by a sonographer. **f**, Bland–Altman plot for the systolic-to-diastolic ratios (S/D ratio) measured by the autonomous tracking algorithm and a sonographer. The solid red line is the mean difference between the two methods, the solid blue lines are 95% limits of agreement (1.96 standard deviations above and below the mean differences), and the black dashed line is the zero difference between the two methods.

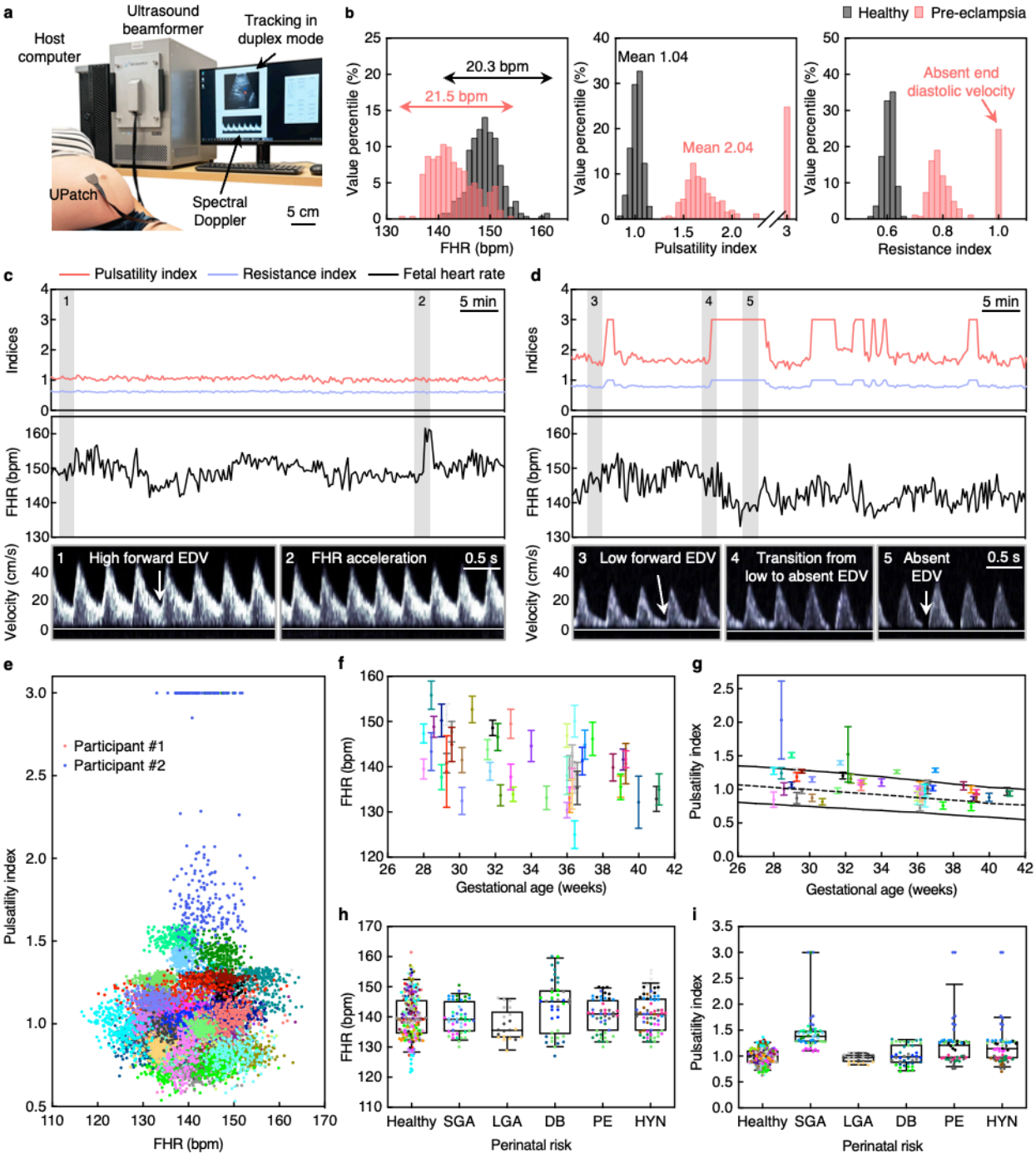

Fig. 4 | Continuous monitoring of pregnant participants. a, Photograph of the measurement setup, with participants in a semi-recumbent position. The UPatch was laminated on the maternal abdomen and connected to an ultrasound beamformer and a host computer, which displays the duplex image, tracks the sample gate, and measures spectral Doppler signals

from the umbilical artery. **b**, Histograms of the fetal heart rate (FHR, left), pulsatility index (middle), and resistance index (right) from the healthy (grey) and pre-eclamptic (red) participants. **c**, Pulsatility index, resistance index (top), and FHR (middle) recording from the healthy participant, with highlighted spectral Doppler signals from the shaded regions (bottom). The recorded traces share the same scale bar. The spectra share the same scale bar. **d**, Pulsatility index, resistance index (top), and FHR (middle) recording from the pre-eclamptic participant, with highlighted spectral Doppler signals from the shaded regions (bottom). The recorded traces share the same scale bar. The spectra share the same scale bar. **e**, Scatterplot of pulsatility index against FHR with each color in the data points corresponding to an individual participant. Participant #1 corresponds to the healthy participant and Participant #2 corresponds to the pre-eclamptic participant in **b** and **c**. The corresponding color scheme for all participants is provided in Supplementary Figs 88–91. **f**, FHR plotted against gestational age (n = 241). **g**, Pulsatility index plotted against gestational age (n = 241). The black dashed line is the 50^{th} percentile, and the solid black lines are the 5^{th} and 95^{th} percentiles of a widely used reference population[23]. Data in **f** and **g** are presented as mean ± standard deviations. **h**,**i**, Box plots of (**h**) FHR and (**i**) pulsatility index stratified by perinatal conditions: healthy (31 participants, 217 data points), small for gestational age (SGA; 7 participants, 49 data points), large for gestational age (LGA; 3 participants, 21 data points), diabetes (DB; 6 participants, 42 data points), pre-eclampsia (PE; 7 participants, 49 data points), and maternal hypertension (HYN; 10 participants, 70 data points). Each box represents the interquartile range (25^{th} to 75^{th} percentiles), the whiskers denote the 5^{th} to 95^{th} percentile range, and the midline indicates the median. Each participant's data was segmented into 10-min intervals (n = 7 per participant). The color schemes in **e**–**i** are the same.

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Methods

Fabrication of the UPatch

622 The fabrication process can be divided into three steps: (1) transducer assembly, (2)
623 electrode printing and acoustic lens integration, and (3) device packaging.

624 (1) Transducer assembly**:** A silicone (Ecoflex–0030, Smooth–On) layer was first spin-
625 coated at 3000 rpm for 60 s on two glass slides and cured in an 80 ºC oven as temporary
626 substrates. Conductive epoxy (Von Roll 3022 E–Solder, EIS) was prepared as the backing layer
627 by mixing an Ag–epoxy composite with a hardener in a 12.5:1 weight ratio and subsequently
628 degassed in a vacuum desiccator. Then, a 1–3 composite (Del Piezo Specialties) was placed on
629 the silicone surface of one glass slide. The degassed conductive epoxy was placed on top and
630 covered by the other silicone-coated glass slide, followed by curing at room temperature for 24
631 h. The thickness of the backing layer (400 μm in this study) was controlled by placing certain
632 spacers in between the two glass slides.

633 After curing, the glass slides were removed, and the transducer made of 1–3
634 composite/backing was diced. To ensure direct transfer of the backing layer side of the
635 transducer to the signal electrodes, the 1–3 composite side of the transducer was first placed on
636 an ultraviolet tape (UHP–110M3, Denka) with a frame film applicator (UH114, Ultron Systems,
637 Inc.). Then, the transducer was diced with a dicing machine (DAD3221, Disco) to 25.55 mm by
638 12 mm using a diamond blade (ZH14–SD1700–V1–90 GE, Disco) at 25,000 spindle rotations
639 per minute (rpm) and 0.5 mm/s feed speed. A second diamond blade (Z09–SD4000–Y1–60,
640 Disco) was used to dice the 0.05 mm kerf of the transducer. Super multipass dicing was
641 performed with 50 μm depth stepwise cuts at 50,000 spindle rpm and 10 mm/s feed speed to
642 minimize blade and transducer vibration during dicing (Supplementary Fig. 2 and Supplementary
643 Discussion 7). Finally, the diced transducers were detached from the ultraviolet tape by an
644 ultraviolet curing system (UH104, Ultron Systems Inc.).

(2) Electrode printing and acoustic lens integration: The substrate for electrode printing was prepared by spin-coating polydimethylsiloxane (Sylgard 184) at 3000 rpm for 60 s on a glass slide, followed by curing in an 80 ºC oven as a temporary substrate. Then, polyimide/copper laminates (AC181200EN, DuPont) and the polydimethylsiloxane–coated glass slide were activated by ultraviolet light (PSD series Digital UV Ozone System, Novascan) and bonded together. Electrode patterns (such as electromagnetic shielding, mesh ground electrode, and two signal electrodes) designed with AutoCAD (Autodesk) were laser ablated (G4 Pulsed Fiber Laser, wavelength 1095 to 1065 nm, energy 0.228 mJ, frequency 35 kHz, speed 300 mm/s, and pulse width 500 ns) on the polyimide/copper laminate (Supplementary Fig. 9). Two polyimide masks were prepared by laser ablation for dispensing conductive epoxy on the transducer bonding pads of the two signal electrode layers.

For acoustic lens integration, an aluminum block mold was designed with Solidworks (Dassault Systemes) and machined using a computer-numerically controlled machining (Tormach 1100MX, Tormach Inc.) at a spindle speed of 10,000 rpm and feed rate of 40 in/min (Supplementary Fig. 6 and Supplementary Discussion 8). The mold was cleaned with isopropanol before silicon elastomer was cured in the mold and used as the acoustic lens.

(3) Device packaging: First, polyimide tapes were adhered to two glass slides to ensure device detachment after final packaging. Silicone (Ecoflex–0030, Smooth–On) mixed with black dye was spin–coated on the polyimide tape at 4000 rpm for 60 s, followed by curing at room temperature for 2 h. The electromagnetic shielding layer, which was transferred onto a water–soluble tape (AQUASOL), and the silicone–coated glass slide was activated by ultraviolet light and bonded together, followed by heating in an 80 ºC oven for 30 min. After the water–soluble

667 tape was dissolved, a silicone layer was spin–coated on top at 4000 rpm for 60 s followed by
668 curing.

669 Similarly, the first layer of the signal electrode was aligned and transferred onto the
670 electromagnetic shielding layer via a water–soluble tape. A polyimide mask was placed on the
671 signal electrode to protect the transducer bonding pads of the electrode from another silicone
672 layer spin–coated on top. After the polyimide mask was removed and the silicone layer was
673 cured, the second layer of the signal electrode was aligned and stacked onto the first layer of the
674 signal electrode. Conductive epoxy was dispensed onto the transducer bonding pads of both
675 signal electrodes through a polyimide mask. After the mask was removed, the backing layer side
676 of the diced transducers was aligned and bonded by curing the conductive epoxy for 8 h at room
677 temperature and then 2 h at 40 ºC. Flexible printed circuit cables were bonded to the signal
678 electrodes via solder paste ($Sn_{42}Bi_{57.6}Ag_{0.4}$, m.p. 138 ºC).

679 The mesh ground electrode was bonded to a silicone–coated glass slide. After conductive
680 epoxy was dispensed using a polyimide mask, the mesh ground electrode was aligned and cured
681 onto the transducers. The exterior boundaries of the mesh ground electrode and electromagnetic
682 shielding layer were connected with vertical interconnect accesses by soldering copper wires
683 (Remington Industries) to create a Faraday cage (Supplementary Fig. 10 and Supplementary
684 Discussion 9)[76]. After the device was fully encapsulated with silicone elastomer, the glass slides
685 were removed. Finally, the acoustic lens was aligned and bonded with silicone elastomer onto
686 the packaged transducers.

687

688 **Quality assurance**

The UPatch was transmitted at a frequency of 2.5 MHz with a sampling rate of 10 MHz. Apodization was uniform across 64 channels with the imaging view angle of 45°. The number of compounding angles was 21 with a step size of 0.75° for B-mode imaging. The ensemble length was 14 for color flow mode. Doppler modes utilized a pulse repetition frequency of 3000 Hz. The sample gate size for spectral Doppler was 2 mm.

The UPatch was characterized using two phantom setups and compared to a handheld clinical ultrasound device (P4-1, ATL), both transmitting at 2.5 MHz. Both the UPatch and the handheld clinical ultrasound device (P4-1, ATL) were connected to the Vantage 256 (Verasonics), rather than connecting the P4-1 device to a clinical backend system (Voluson E10, GE). This method allows comparison of the probe performance more fairly and avoids confounding effects from differences in backend signal processing.

First, the brightness mode was characterized with a multipurpose phantom (CIRS ATS 539, CIRS Inc.) (Supplementary Fig. 15)[37,38]. Monofilament wires were used to determine the axial and lateral resolutions. A volume of water on top of the phantom was used as a standoff to characterize depths up to 20 cm. The distance between two adjacent pixels in the axial and lateral directions was calculated as:

$$\Delta y = \frac{depth}{N_{pixel,axial} - 1} \quad (1)$$

$$\Delta x = \frac{depth}{N_{pixel,lateral} - 1} \quad (2)$$

The full width at half maximum of the point spread function for each wire in the axial and lateral directions was measured. The axial and lateral resolutions were calculated by multiplying the number of pixels within the full width at half maximum and adjacent pixel

distance. The six tissue–mimicking contrast targets in the phantom were used to characterize the dynamic range and contrast-to-noise ratio[38]. The dynamic range was calculated by linearly fitting among the six greyscale targets and finding the maximum positive and negative contrasts corresponding to grey values of 255 and 0. Then, the negative one was subtracted from the other. The contrast-to-noise ratio was calculated by:

$$Contrast - to - noise\ ratio\ = \frac{|\mu_{in} - \mu_{out}|}{\sqrt{\sigma_{in}^2 + \sigma_{out}^2}} \quad (3)$$

where μ_{in} and σ_{in} are the mean and standard deviation of pixel intensities within each target, and μ_{out} and σ_{out} are the mean and standard deviation of pixel intensities outside each target. The contrast and contrast-to-noise ratio were computed based on displayed images rather than raw radiofrequency signals or in-phase and quadrature (I/Q) data, meaning that the values were derived from log-compressed images[77].

The color and spectral Doppler modes were characterized with Doppler fluid (769DF, CIRS Inc.) pumped into a Doppler phantom (CIRS ATS 523A, CIRS Inc.) with 2, 4, 6, and 8 mm diameter using a peristaltic pump (Supplementary Fig. 16)[39]. The UPatch was positioned at varying distances from 30 mm to 120 mm at intervals of 10 mm from the phantom. The signal-to-noise ratio in the color flow images was calculated by increasing the color gain to maximum to obtain a noise level in the tissue mimicking phantom. The accuracy of spectral Doppler was measured by comparison to a known velocity from the peristaltic pump at different depths and velocities.

Acoustic exposure and thermal measurement protocols

(1) Hydrophone: A needle hydrophone (HNP–0400, ONDA) was connected to a right-angle adapter (ONDA), pre-amplifier (AH–2010–100, ONDA), and DC block (BNP, ONDA) and suspended in deionized water (Supplementary Fig. 24)[78]. The signals were read with an oscilloscope (PicoScope 5000 Series, Pico Technology, Cambridgeshire). At 2.5 MHz, the effective sensitive element diameter (a more meaningful measure of spatial resolution than the geometrical sensitive element diameter, which was 400 μm) was estimated to be 480 μm[79]. The uncertainties of hydrophone-based pressure, intensity, and power measurements are ~ 15%[80].

The hydrophone system was calibrated < 1 year before the measurement. The calibration $M_c(f)$ is the hydrophone end–of–cable open circuit sensitivity, which can be converted by:

$$M_C(f)_{V/Pa} = 10^{\frac{M_C(f)_{dB\ re.1V/\mu Pa}+120}{20}} \quad (4)$$

where $M_C(f)_{V/Pa}$ is the sensitivity in volts per pascal and $M_C(f)_{dB\ re.1V/\mu Pa}$ is the sensitivity in decibels relative to 1 volt per micropascals[78]. The manufacturer's sensitivity magnitude data (i.e., relative calibration from 1 to 20 MHz) was used to perform magnitude–based waveform deconvolution to compensate for the hydrophone's nonuniform frequency response in the signal spectrum (Supplementary Fig. 24). Magnitude–based deconvolution has been shown to be nearly as effective as complex deconvolution for improving the accuracy of pressure and intensity measurements[81]. Additionally, because the ratio of the hydrophone sensitive element diameter d_g to the product of the wavelength λ and $F/\#$ (ratio of focal distance to aperture width) << 1, a hydrophone spatial averaging correction was not necessary[82].

Considering the capacitive load of the preamplifier when converting the voltage measurements to pressure values, the loaded sensitivity $M_L(f)$ can be estimated as:

$$M_L(f) = G(f)M_C(f)_{V/Pa}\frac{C_H}{C_H + C_A + C_c} \quad (5)$$

where $G(f)$ is the preamplifier gain (20 dB in this study), C_H is the capacitance of hydrophone (70 pF in this study), C_A is the capacitance of preamplifier (7 pF in this study), and C_C is the capacitance of right–angle adapter (1.6 pF in this study)[78].

A hydrophone controlled by a 3D linear motor in a water tank was used to measure the acoustic signal intensity (Supplementary Fig. 24). Transverse scans with scanning increments of $\Delta x = 0.5$ mm (approximately one effective hydrophone diameter) and $\Delta y = 0.5$ mm and an axial scan with scanning increments of $\Delta z = 0.5$ mm from break–point depth to focal depth were performed (Supplementary Discussion 11)[83]. Two different beamforming methods were measured separately: diverging beam (for brightness and color flow mode) and focused beam (for spectral Doppler mode) that was focused at every 1 cm depth (between 5 to 15 cm) along the beam axis. The UPatch was activated at 30 V with a pulse repetition frequency of 3000 Hz. Power was derived from the hydrophone by integrating over a transverse plane within the beam cross–sectional area where the intensity was at least –26.2 dB of the peak value. All measurements were calculated based on the settings used for duplex imaging and spectral Doppler because these were the modes utilized during continuous monitoring in this study.

(2) Radiation force balance: Power measurements were obtained from a radiation force balance (RFB–2000, ONDA, Sunnyvale, CA) with a flat absorbing target, which could accurately measure power levels up to 2 W (Supplementary Fig. 25)[84]. First, a 1 g weight was used to calibrate the radiation force balance. The calibration constant obtained was 0.9699 A/N, which fell in the range designed by ONDA, 0.9–1.1 A/N. The distance between the radiation force balance and the UPatch was set to ≥ 1 cm to avoid thermal and electrical coupling effects

between the radiation force balance and the UPatch. Acoustic output power was obtained from the entire aperture of the UPatch, and the bounded–square output power over a 1 cm^2 area was measured by masking the UPatch.

The thermal index was calculated based on acoustic output power measurements from both the radiation force balance and hydrophone scan in the transverse plane, providing two independent measurements to minimize uncertainty in the calculation[46].

(3) Surface temperature testing: The surface temperature of the UPatch was evaluated under two scenarios to simulate clinical conditions based on IEC 60601-2-37 (ref. [85]). Although the standard requires only 30-min testing[85], we extended the assessments to 48 h to ensure compliance with regulatory standards even under prolonged operation.

First, a phantom test was performed to replicate the setup in the hospital (Supplementary Fig. 23). The phantom consisted of a silicone-based skin (NPL, UK) and agar-based tissue (NPL, UK) with acoustic and thermal properties similar to those of human skin and tissue, respectively. A type K thermocouple (5SRTC-TT-KI-40-1M, OMEGA, Norwalk, Connecticut) with a thickness of 75 μm (<< ultrasound wavelength at 2.5 MHz) was placed on top of the skin mimicking phantom to measure the surface temperature of the UPatch[45].

Second, to evaluate thermal changes when the UPatch was activated in air, surface temperature was measured. Because the UPatch/air interface has a higher acoustic impedance mismatch compared to that of the UPatch/tissue phantom interface, more heat is expected to be generated in the air[45]. The UPatch was suspended on a clean surface in a stationary position with minimal airflow across the transducer area. The same thermocouple was adhered to the UPatch for temperature measurement.

796

797 **Evaluation of the vessel–tracking algorithm**

798 The comparative analysis of the lateral and axial discrepancies in defining the sample
799 gates between the tracking algorithm and a sonographer was performed using double–blind
800 method (Supplementary Discussion 18). The tracking algorithm and the sonographer each
801 selected sample gates independently from 1,000 duplex images, without access to the other's
802 chosen locations. This design was used to prevent bias in sample gate placement and to ensure an
803 objective comparison between algorithm-selected and sonographer-selected measurements, with
804 discrepancies quantified by the spatial distance between the selected gates. The lateral and axial
805 discrepancies were calculated between the two sample gates (Fig. 3d). A 2 mm sample gate
806 length is commonly employed in hospitals (Supplementary Discussion 19)[1]. Thus, within a 2 mm
807 discrepancy between the tracking algorithm and the sonographer, the sample gate was considered
808 to acquire spectral Doppler signals accurately.

809 Additionally, a new dataset was acquired, containing 500 duplex images from scanning
810 the clinical device on a participant and 500 duplex images from the UPatch on another
811 participant. The tracking algorithm identified the sample gates in all 1,000 images. We engaged
812 three experienced sonographers, each with 30, 8, and 20 years of fetal ultrasonography
813 experience, to independently assess the tracking performance on 1,000 images (Supplementary
814 Fig. 39). This validation method aligns with the recently FDA-cleared cardiac Doppler tracking
815 algorithm[86], which also relied on three sonographers and evaluated only 168 images. All three
816 sonographers evaluated the tracking algorithm with a customized graphical user interface in
817 Matlab R2023b (Mathworks). "Optimal" was defined by the sonographer as the tracked sample

gate that could potentially be used for spectral Doppler acquisition, “Sub-optimal” was defined as the sample gate potentially providing low-quality spectral Doppler signals, and “Non-diagnostic” was defined as the sample gate that could not provide any spectral Doppler signal (Supplementary Fig. 39 and Supplementary Discussion 18).

Finally, the tracking algorithm was evaluated with the UPatch based on the systolic-to-diastolic ratio of a participant (Supplementary Discussion 18). Each pair of measurements was acquired sequentially by a sonographer and the tracking algorithm. For the mean difference ± standard deviation between the UPatch and clinical device, it is expected that the device differences and physiological variability increase the spread of values, even when the overall bias remains very small (Fig. 2f). By eliminating inter-device variability and focusing merely on sample gate selection, the tracking algorithm analysis yields a narrower distribution, even though the mean offset is slightly larger (Fig. 3f). Notably, the mean values reflect different reference comparisons and are not expected to be identical, whereas the reduced standard deviation reflects the elimination of device-related variability.

Human test protocol

All human tests were approved by the Institutional Review Board of the University of California San Diego (#804817) and the UK Research Ethics Committee (23/WA/0032). The participants all gave voluntary consent to the UPatch tests, and all measurements were conducted by a clinician. The study included 62 participants for accuracy validation (Fig. 2f, Table 1, and Supplementary Fig. 32) and 52 participants for continuous monitoring (Supplementary Figs 40–95). This dataset captures a broad range of clinical conditions across the study cohort, including

variations in fetal biometry (e.g., small and large for gestational age) (Fig. 4e–i and Supplementary Figs 96 and 97) and gestational age (Table 1). For the validation test, three pairs of measurements were acquired from the UPatch and a handheld clinical ultrasound device (Voluson E10, GE) on 62 participants. For continuous monitoring, the clinical device was used to locate the position of the vessel. Then, the UPatch was taped on the maternal abdomen for continuous monitoring. The current study was conducted under inpatient conditions, where pregnant participants were free to move within the limits of standard clinical monitoring. During the recordings, the pregnant participants naturally moved within their beds, accompanied by various fetal behavioral states, fetal movements, and uterine contractions.

While healthy pregnancies typically exhibit strong forward end diastolic flow, severe pathological pregnancies can have reduced forward end diastolic flow, absent end diastolic flow, or reversed end diastolic flow, resulting in weaker diastolic spectral content and increased waveform variability[3,87]. In these cases, pulse repetition frequency of Doppler ultrasound becomes particularly critical: if it is too high, low velocity components are compressed near the waveform baseline and become difficult to distinguish from noise, while a pulse repetition frequency that is too low risks aliasing at peak systolic velocities[33]. To ensure a fair and clinically relevant comparison across healthy and severe pathological conditions, validation was performed using matched pulse repetition frequency settings between the UPatch and clinical ultrasound system.

The UPatch was designed for easy repositioning as needed. It is movable, can be sanitized with alcohol wipes, and can be repositioned without compromising functionality. This allows adaptability in clinical use while maintaining hygiene and patient comfort.

For fetal biometry, estimated fetal weight (EFW) was calculated based on Hadlock's IV formula[49]:

$$EFW = 1.3596 - 0.00386AC \cdot FL + 0.0064HC + 0.00061BPD \cdot AC + 0.0424AC + 0.174FL \quad (6)$$

where AC is abdominal circumference, FL is femur length, HC is head circumference, and BPD is biparietal diameter.

Data availability

The data supporting the findings of this study are available in the main text or the Supplementary Information.

Code availability

The code used in this study is available from GitHub (https://github.com/ParkingTom/Autonomous-tracking-algorithm).

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Additional information

Supplementary Information is available for this paper.

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Table of Contents

Supplementary Discussion 1: Continuous prenatal monitoring

The primary goal of antepartum fetal surveillance (also referred to as prenatal monitoring) is to prevent irreversible fetal injury or stillbirth[1]. Current methods, including cardiotocography[2], fetal movement[3], and ultrasound-based assessments[4], are performed intermittently and seek to detect subacute uteroplacental insufficiency[1,5]. These methods focus on monitoring the gradual changes in fetal health, allowing healthcare providers to intervene before any permanent damage occurs. However, the intermittent nature of these assessments means that they may miss sudden, acute events that can lead to fetal compromise.

Many stillbirths are suspected to result from acute events, such as umbilical cord accidents, placental abruption, and premature rupture of the membrane[5]. These acute events cannot be predicted by the current intermittent approaches. Umbilical cord accidents, including cord compression, nuchal cord (where the cord is wrapped around the baby's neck), and prolapsed cord (where the cord slips into the birth canal ahead of the baby), can lead to a rapid decrease in the fetal oxygen supply[6]. Placental abruption is a serious pregnancy complication in which the placenta partially or completely detaches from the inner wall of the uterus[7]. This detachment can disrupt the supply of oxygen and nutrients to the fetus and cause severe internal bleeding in the uterus[7]. Premature rupture of the membrane (also known as amniotic sac) can lead to infections and preterm labor, posing high risks to both the mother and fetus[8].

Continuous prenatal monitoring can detect both subacute and acute fetal compromise. It has the potential to identify signs of sudden fetal distress as they occur, allowing for timely interventions that could prevent stillbirth and other severe outcomes.

Supplementary Discussion 2: Wearable fetal monitors

Numerous wearable devices have emerged to extract fetal signals in combination with other maternal sensors (Supplementary Fig. 1)[9-11]. Fetal monitoring technologies vary considerably in terms of sensing modality, number and configuration of sensors, spatial resolution, and clinical application. These wearable devices can continuously detect fetal heart rate, electrocardiogram, oximetry, and movements. Here, we provide a summary of state-of-the-art technologies, highlighting general limitations. Additional technical details for these technologies can be found in the cited references.

First, wearable fetal heart rate monitors, such as HeraBEAT (HeraMED)[12] and Moyo fetal heart rate monitor (Moyo)[13], have been introduced to address the limitations of handheld devices, such as Sonoline B (Sonoline)[14], SD1 (Edan)[15], and Newman (Babybeat)[16] that only offer intermittent measurements. These devices utilize ultrasound and operate in continuous wave Doppler mode, leveraging two ultrasound transducers for transceiving[9]. Although Doppler signals are intended to originate from the movements of fetal heart valves, any movements are translated into signals[9]. The absence of spatial resolution of the continuous wave Doppler mode is a crucial limitation because it results in the mixing of different signals. Additionally, during advanced gestation, the increased frequency of uterine contractions can influence fetal heart rate[17]. Consequently, the measured fetal heart rate may not be accurate, and therefore, monitoring fetal heart rate alone does not adequately reflect the physiological changes within the uterus.

To better understand how uterine contractions influence fetal heart rate, wearable cardiotocography can simultaneously monitor these two signals. Traditional wearable cardiotocography, such as Corometrics 250cx (GE Healthcare)[18], Avalon FM50 (Philips)[19], and Sense4baby (Advanced Maternity Innovations)[20], relies on continuous wave Doppler for fetal heart rate and a tocodynamometer for uterine contractions (Supplementary Fig. 1). Due to their large form factor, these devices necessitate belts for optimal sensor placement and wired connections to external hardware for power and signal transfer. Such fixations impose constraints on the maternal abdomen, making them less feasible for prolonged use. Additionally, the mechanical mismatch between the bulky devices and the maternal skin is prone to signal loss and motion artifacts.

Minimally invasive methods, such as fetal scalp electrodes and intrauterine pressure catheters, provide more direct measurements of fetal heart rate and uterine contractions, enhancing signal integrity under critical conditions[17]. Accurate monitoring becomes essential in high-risk pregnancies to enable prompt interventions because fetal and maternal movements can disrupt cardiotocography signals[21]. Additionally, amniotic sac rupture may reduce fluid levels, hindering ultrasound transmission[17]. Scalp electrode provides direct fetal heart rate tracings by inserting spiral wires into the fetal head. Intrauterine pressure catheters quantify the pressure intensity of the uterine contraction, while tocodynamometers only monitor the contraction frequency and duration[22]. However, the use of scalp electrodes and intrauterine catheters is restricted to labor monitoring[17]. In addition, scalp electrodes may lead to complications such as scalp trauma[17], while intrauterine pressure catheters may cause infections and uterine rupture[22].

More recently developed devices have focused on miniaturized and flexible wearable cardiotocography, such as ANNE (Sibel Health)[23] and Farus wearable ultrasound and tocodynamometer (Farus)[24], which minimize the mechanical mismatch between the device and the maternal skin (Supplementary Fig. 1). These devices integrate power, signal processors, and wireless connections into the device casing, eliminating the need for securing belts. This advancement largely improves portability and user-friendliness in wearable cardiotocography.

Second, wearable electrocardiogram devices employ biopotential electrodes, such as AN24 (Monica Healthcare, now Novii by GE Healthcare)[25,26], Avalon beltless fetal monitoring solution (Philips)[27], Invu (Nuvo)[28], Meridian M110 (MindChild Medical)[29], Bloomlife electrophysiological monitor (Bloomlife)[30], Nemo fetal monitoring system (Nemo Healthcare)[31], and Femom lite (Biorithm)[32]. Maternal electrocardiograms (~200 μV) often interfere with weak fetal electrocardiograms (~3 μV)[25]. Additionally, movement and positional changes of the fetus cause fluctuations in electrical impedance, further compromising signal integrity[25]. To overcome the low signal amplitude of fetal electrocardiograms, various systems utilize an array of electrodes on the maternal abdomen with advanced signal decomposition algorithms to isolate the subtle fetal from the maternal electrocardiograms. The AN24 uses a five-electrode configuration: one as a common ground and the remaining four placed in upper, lower, right lateral, and left lateral positions, achieving 85% reliability in estimating fetal heart rate (Supplementary Fig. 1)[26]. The Meridian M110 (MindChild Medical) uses 32 electrodes distributed across the maternal abdomen and has demonstrated 89.9% reliability[33]. While the increased electrode density enhances spatial resolution to handle fetal movement and signal interference, the modest improvement in reliability suggests that increasing the electrode count adds complexity to circuitry and signal processing but offers diminishing marginal benefits.

Third, wearable fetal oximetry devices, such as Lumerah (Raydiant Oximetry)[34] and Storx intrapartum fetal monitoring (Storx Technologies)[35], have been proposed to measure fetal blood oxygen levels. However, light penetration through the various tissue layers poses immense challenges, especially with signal interference from maternal signals and motion artifacts[36]. For example, while the initial placement of a wearable oximetry device can be optimized, fetal movements may cause signal loss. Consequently, reliably measuring blood oxygen levels in human fetuses has not yet been demonstrated.

Finally, wearable fetal movement devices are based on accelerometers (Supplementary Fig. 1)[37-42]. Accelerometers can capture various signals, including both fetal and maternal movements, which are often irregular[38]. Additionally, when the position of the fetus changes, the signal amplitude of different fetal movements also changes[42]. Therefore, it is challenging to distinguish which signals are fetal movements. To improve signal quality, the integration of acoustic sensors with accelerometers has been explored[41,42]. These devices are currently in the proof-of-concept phase, with belts and sleeves employed to secure the accelerometers, signal processing units, and batteries.

Supplementary Discussion 3: Implications and limitations of cardiotocography

Cardiotocography measures fetal heart rate and uterine contractions[9]. Changes in baseline, variability, and transient accelerations and decelerations of the fetal heart rate are due to the fetal central nervous system's responses to various environmental changes[17].

Fetal heart rate and heart rate variability serve as important indicators of fetal well-being[9,17,43]. These parameters reflect the physiological responses of the fetus under varying conditions, particularly the oxygenation state. Thus, monitoring these parameters has been widely used for identifying conditions arising from deprived oxygen levels, such as impending asphyxia during labor[44].

Typically, the fetal heart rate lies within the range of 120–160 beats per minute (bpm), with the baseline declining with advanced gestation[9]. Deviations from this range include bradycardia, defined as a baseline below 120 bpm, and tachycardia, characterized as a baseline exceeding 160 bpm[43]. Variability is categorized as absent, minimal (1 to 5 bpm), moderate (5 to 25 bpm), or marked (>25 bpm)[43]. Moderate and marked variabilities are often regarded as normal physiological responses as the sympathetic and parasympathetic nervous systems adjust the fetal heart rate based on dynamic in-utero conditions[43]. Absent or minimal variabilities may indicate hypoxia but are not always reliable as they may also be associated with other factors such as fetal sleep cycle[9], and thus need further evaluation[43].

Acceleration (transient increase in the fetal heart rate) suggests optimal fetal well-being[17]. Such elevation is often associated with fetal movements or external stimuli. Importantly, regular occurrences of accelerations are a reassuring sign, indicating that the fetus is well-oxygenated and the nervous system is responsive to its movements and external stimuli[17].

Uterine contractions are used in monitoring the onset of active labor as the uterus and cervix prepare for labor[22]. The tocodynamometer measures the frequency and duration of uterine contractions by placing a pressure transducer on the maternal abdomen to sense and record changes in tension as the uterus contracts[22].

The uterus, composed of smooth muscles, is capable of progressive stretching[17]. In the antepartum period, two primary types of contractions are observed[17]. First, sporadic and weak contractions of short duration occur in a localized area in the uterus. These contractions often occur in early stages of pregnancy[17]. Second, Braxton Hicks contractions, which are more intense, may take place irregularly, approximately once per hour at around 30-week of gestation, increasing up to every 5 to 10 min at term[17]. More frequent and regular contractions signal the onset of labor, leading to cervical dilation[17]. During labor, uterine contraction monitoring is used to detect labor abnormalities. For example, hypocontractile or hypercontractile uterine activity can lead to prolonged labor or uterine tachysystole, respectively, posing risks to fetal health[17].

By using cardiotocography during labor, the timing between fetal heart rate and uterine contraction has been associated with fetal health[17]. The fetal heart rate response to uterine contractions can indicate how well the fetus is coping with external stress[17]. Specifically, these patterns are categorized into three distinct types (i.e., early, late, and variable decelerations), each defined based on the temporal relationship between fetal heart rate and uterine contractions[43].

Early deceleration is characterized by a concurrent decline in fetal heart rate with the onset of uterine contraction[43]. This phenomenon is a result of pressure exerted on the fetal head during contraction, which in turn alters cerebral blood flow[17]. Then, the central nervous system is stimulated, prompting a deceleration in fetal heart rate. As the contraction amplifies, the fetal heart rate continues to decelerate. Finally, as contraction subsides, the fetal heart rate returns to the baseline. Early deceleration demonstrates the adaptive capabilities of the fetal central nervous system and is not indicative of fetal distress[43].

Late deceleration is characterized by a subsequent decline in fetal heart rate following the onset of uterine contraction[43]. Specifically, the nadir of the fetal heart rate is observed post-peak contraction. This deceleration process is commonly attributed to transient fetal hypoxia due to placental insufficiency[17]. In scenarios where placental function is sub-optimal, diminished blood flow during contractions may result in reduced oxygenation from the placenta to the fetus[17]. In response to the low oxygenation state, the fetal central nervous system decelerates the heart rate. As contraction ends, the fetal heart rate begins to recover and reverts to its baseline heart rate.

Variable deceleration is characterized by the unpredictable onset of fetal heart rate relative to the timing of uterine contractions[43]. These decelerations exhibit an abrupt onset and decline without a consistent relationship to the contraction phase. The predominant factor for variable decelerations is umbilical cord compression, leading to a transient decrease in blood flow between the fetus and the placenta[17]. This hemodynamic alteration elicits a temporary deceleration in fetal heart rate. As the umbilical cord blood flow returns to normal, the fetal heart rate returns back to the baseline.

While cardiotocography is useful, monitoring merely the fetal heart rate and uterine contraction comes with inherent challenges and has led to misdiagnosis[9,45]. Consequently, cardiotocography cannot serve as a stand-alone diagnostic tool and requires additional sensors for a more comprehensive and accurate assessment of fetal well-being.

First, cardiotocography interprets fetal well-being based on the central nervous system's response to alterations in blood flow[9,17]. Thus, the reliability of these measurements is constrained by the complexity of the underlying biological processes. This means that these measurements indirectly reflect fetal well-being, serving more as a potential consequence rather than a direct cause of the fetal health.

Second, alterations in fetal heart rate are one of the many central nervous responses. Others include systemic vasoconstriction and vasodilation of the intricate cardiovascular system[46]. In severe cases, the brain–sparing effect may be activated, resulting in the dilation of cerebral vessels to enhance brain perfusion, while vessels in other areas are constricted[46].

Third, the intrinsic variability in fetal heart rate presents challenges in deciphering the significance of specific patterns[9]. Determining the threshold for clinical intervention requires distinguishing between transient benign changes and those that are indicative of sustained fetal compromise. For instance, growth-restricted fetuses can sometimes exhibit a normal fetal heart rate range[17] and variability[2], while fetuses with bradycardia can be found to be well-oxygenated post-delivery[45]. Such data can precipitate unnecessary interventions, including Cesarean sections or operative vaginal deliveries, thereby exposing both the mother and the fetus to additional risks, including potential postnatal complications[9].

Consequently, cardiotocography has a false-positive rate of up to ~50% (ref. [9]), leading to unnecessary interventions and preterm birth[2,47]. While cardiotocography is a valuable source of information, findings must be integrated within a broader clinical context, limiting the effectiveness of cardiotocography as a stand-alone diagnostic approach.

Supplementary Discussion 4: Blood flow monitoring with ultrasonography

Ultrasonography is an invaluable tool in contemporary obstetrics due to its safety and diagnostic effectiveness[48-51]. Unlike fetal heart rate monitoring, which measures the response of the fetal central nervous system to alterations in blood flow (Supplementary Discussion 3), spectral Doppler ultrasound provides a direct quantitative measurement of the fetal and placental blood flow[1,52].

First, prenatal diagnostic options are notably limited compared to those available postnatally. Postnatal diagnostics utilize direct observations (such as skin paleness, scarring, and coughing), pain symptoms, and various measurements (such as blood pressure, body temperature, and oxygen saturation). In comparison, prenatal assessments are based on maternal perceptions and ultrasound-based measurements[3,53]. Additionally, while postnatal diagnoses often use blood pressure measurements as a standard indicator of cardiovascular well-being, prenatal monitoring heavily relies on ultrasound-based blood flow assessments[48].

Second, blood flow is generally stable in healthy fetuses[54]. The intrinsic variability in fetal heart rate caused immense challenges for accurate diagnosis with cardiotocography (Supplementary Discussion 3). Parameters such as umbilical artery blood flow exhibit minimal variations even during maternal physical exercise[55]. The relative stability of blood flow suggests that alterations are indicative of hypoxia or other vulnerable states in the fetus[55].

Third, fetal hypoxia is a primary cause of adverse prenatal and perinatal outcomes, including intrauterine growth restrictions[49,50], stillbirths[56], and hypoxic-ischemic encephalopathy[57], which may result in cerebral palsy[58] and permanent disability[52]. Spectral Doppler is an essential tool to diagnose hypoxic fetuses and aids in distinguishing pathological from healthy conditions[59-62]. For example, fetal biometry, while instrumental in detecting small for gestational age fetuses, does not unequivocally differentiate intrauterine growth restriction from small yet healthy fetuses[61]. Umbilical artery Doppler signals effectively differentiate the two cases. Intrauterine growth restriction is often associated with abnormal blood flow patterns (such as absent or reversed end

diastolic flow), while small but healthy fetuses are characterized by normal blood flow patterns (forward end diastolic flow)[49,50].

Fourth, third-trimester pregnancy is often the most vulnerable when the fetal demand for oxygen and nutrients reaches its peak and there is a higher tendency for cord accidents (Supplementary Discussion 1)[63]. With prolonged distress, fetal asphyxia is the final common pathway leading to stillbirth[64]. Thus, blood flow assessment can guide the timing of delivery complicated by fetal hypoxia[49,51]. Profound compromise in blood flow may necessitate earlier delivery to prevent irreversible organ damage or stillbirth. During such critical states, the brain-sparing effect, a cerebral autoregulation mechanism to continuously supply sufficient oxygen to the brain, may be activated by the vasodilation of cerebral vessels and vasoconstriction of other vessels[46,65]. The surveillance of fetal blood flow allows for the optimization of the timing of delivery, balancing the risks of prematurity against the risks of hypoxia[49].

Despite the importance of continuous blood flow monitoring, current clinical protocols only provide intermittent ultrasound-based measurements in high-risk pregnancies[49,50]. This intermittent approach may overlook emergent complications or severe fetal distress, potentially increasing the risk of stillbirth. Continuous blood flow monitoring via spectral Doppler can provide real-time information about fetal well-being, enable clinicians to make informed decisions, and eventually reduce the likelihood of complications associated with undiagnosed fetal distress[48,50,52].

Supplementary Discussion 5: Clinically relevant vessels for monitoring

In fetal ultrasonography, three indices derived from spectral Doppler waveforms are used to assess fetal well-being based on reported reference ranges[66,67].

The systolic-to-diastolic ratio ($S/D\ ratio$) is defined as:

$$S/D\ ratio = \frac{PSV}{EDV} \tag{1}$$

where PSV is the peak systolic velocity and EDV is the end diastolic velocity.

The pulsatility index (PI) is defined as:

$$PI = \frac{PSV - EDV}{MFV} \tag{2}$$

where MFV is the mean flow velocity.

The resistance index (RI) is defined as:

$$RI = \frac{PSV - EDV}{PSV} \tag{3}$$

Routine hospital examinations of spectral Doppler indices are conducted in three major vessels: umbilical artery, umbilical vein, and middle cerebral artery[60].

The umbilical cord is a helical free-floating tube composed of two arteries and one vein. The umbilical arteries transport deoxygenated blood from the fetus to the placenta[50]. The spectral Doppler signal exhibits low resistance and continuous forward flow throughout systole and diastole in a healthy fetus[50]. Additionally, resistance decreases (i.e., increasing end diastolic

velocity) from the fetal insertion point to the placental insertion point along the length of the umbilical cord, because the branching of the umbilical artery at the placental end results in a larger cumulative cross-sectional area and greater compliance of the vessels[49,50]. In healthy pregnancy, the $S/D\ ratio$, PI, and RI decrease with gestational progression[50]. As a reference, the median $S/D\ ratio$ diminishes from 3.23 at 24-week to 2.68 at 32-week and 2.19 at 40-week gestation; the median PI diminishes from 1.10 at 24-week to 0.96 at 32-week and 0.79 at 40-week gestation; and the median RI diminishes from 0.69 at 24-week to 0.63 at 32-week and 0.55 at 40-week gestation[68]. However, a substantial increase in these Doppler indices or the critical finding of absent or reversed end diastolic flow can signify fetal growth restriction[49,50].

The umbilical vein delivers oxygenated blood from the placenta to the fetus, typically showing continuous and non-pulsatile flow[50]. Fetal breathing is commonly observed in the umbilical venous flow in the third trimester due to changes in intrathoracic pressure[50]. On the other hand, pulsatile umbilical vein waveforms, which reflect the pulsatility of the fetal heart rate, can indicate a detrimental defect in fetal circulation, especially in the fetal heart and placental function[49,50]. A compromised fetal heart can cause back pressure through the fetal right ventricle, right atrium, ductus venosus, and into the umbilical vein[50]. Placental insufficiency can also alter the normal pressure gradient in the umbilical vein[50].

The middle cerebral artery is a major branch of the circle of Willis, carrying oxygenated blood for the fetal cerebral circulation[49]. Similar to the umbilical artery waveform, the middle cerebral artery spectral Doppler shows pulsatile forward flow throughout the cardiac cycle under normal conditions[50]. Two abnormalities can be monitored. First, the brain-sparing reflex is a physiological response to compensate for hypoxic or nutrient-deficient conditions[49]. Blood flow redistributes to prioritize the brain for survival and development[49]. This is characterized by increased end diastolic velocity in the spectral Doppler of the middle cerebral artery, which is reflected by a low PI[49]. Clinically, the cerebroplacental ratio (CPR), defined as the PI ratio of the middle cerebral artery to the umbilical artery, serves as an index to gauge the adequacy of fetal brain perfusion relative to placental blood flow[50]. The median CPR is 1.37 at 20-week, 2.34 at 30-week, and 1.80 at 40-week gestation[46]. Second, fetal anemia is characterized by decreased circulating red blood cells in the fetus[51]. An increased PSV in the middle cerebral artery, exceeding 1.5 times the median at a particular gestational age, is indicative of fetal anemia[51]. The median PSV is 25.5 cm/s at 20-week, 40.5 cm/s at 30-week, and 64.4 cm/s at 40-week gestation[69].

Supplementary Discussion 6: Novelty of the UPatch

The UPatch enables continuous fetal blood flow monitoring, which is essential for high-risk pregnancies.

A key limitation of current ultrasound-based fetal blood flow assessments is the need for a trained sonographer to manually place the sample gate, making continuous monitoring impractical[70]. The autonomous vessel tracking algorithm demonstrated in this work eliminates this limitation, enabling continuous measurements even during fetal movement. This advancement is particularly valuable in low-resource settings, where access to skilled sonographers is often limited or unavailable[71].

Previous wearable ultrasound patches developed for cardiac[72] and vascular[73] imaging employed relatively small apertures at 12.8 mm and 12 mm, respectively, which were sufficient for imaging large cardiac structures and adult vessels at relatively shallow depths. In contrast, the UPatch is designed to image substantially deeper and smaller fetal vessels. To accommodate these requirements, the UPatch aperture was increased to 25.6 mm, representing more than a twofold increase and enabling much enhanced sensitivity and penetration for deep fetal imaging (Extended Data Fig. 1).

The previous design did not incorporate elevational focusing, relying instead on unfocused elevational beam profiles[72]. In contrast, the UPatch integrates an acoustic lens that enhances elevational focusing, reducing the elevational focal beamwidth by 25% and shifting the elevational focal depth from 14 cm to 10 cm. This design enables more balanced elevational resolution across the ~20 cm imaging depth of a typical womb[74].

Conventional dicing strategies in the previous designs limited the achievable kerf width to 0.1–0.2 mm, constraining element density and thereby limiting signal-to-noise ratio[72]. In contrast, the use of super multipass dicing approach reduced the kerf width to 0.05 mm, increasing element width by >50% and improving signal-to-noise ratio while maintaining a minimal level of crosstalk between adjacent elements.

Electromagnetic shielding in prior designs used two isolated electrode layers of liquid metal[72] or copper mesh[73], which provided two-dimensional shielding but could not block electromagnetic interference from entering through the edges of the two electrodes[75]. In contrast, the UPatch was designed with a soft Faraday cage that was made of copper mesh and stitching vertical interconnect accesses. The soft Faraday cage connects the edges of the two electrodes using stitching vertical interconnect accesses. This design enables a three-dimensional soft Faraday cage, providing more effective electromagnetic shielding than prior planar approaches and improving signal integrity across various conditions encountered during wearable operation.

Supplementary Discussion 7: Transducer dicing

Transducer dicing is a critical fabrication step in ultrasound array design. There are two primary dicing methods that are utilized: laser-based and blade-based dicing.

For transducer dicing using a laser, a key challenge is the thickness of the entire material stack. The UPatch is composed of PZT-5H 1–3 composite and conductive epoxy backing layer (Von Roll 3022 E–Solder, EIS) with a total thickness of ~1 mm. First, while laser dicing can effectively cut through PZT, it is limited to thin material stacks (e.g., 0.262 mm[76], 0.08 mm[77], and 0.045 mm[78]). As a result, laser dicing is feasible for selectively dicing the PZT layer only or for high-frequency applications where the entire material stack is very thin. However, it is impractical for achieving precise dicing of the 1 mm-thick material stack used in this study. Second, a key limitation of laser dicing is the heat generated during processing[77], which evaporates Pb to reduce its concentration and possibly induces a metastable pyrochlore phase, leading to a complete loss of piezoelectric properties[79]. Annealing and repolarization are required to restore piezoelectric properties, introducing additional processing steps.

Dicing ultrasound transducers typically utilizes blades with diamond grits embedded in electroplated nickel binders[80]. As the blade rotates, the diamond grits abrade the transducers. The

sharp edges of the diamond grits mechanically chip away microscopic fragments of the transducers[80]. However, dicing thick material stacks with conventional mechanical dicing techniques remains challenging. To address this, additional preparation steps are often required to separately process the PZT and backing layer[81,82]. A common approach involves interposers, which are 3D-printed with the target pitch[83]. The backing layer is molded within the interposers, which separate individual channels. The PZT is then placed on top of the backing-interposer structure with careful alignment. Finally, the PZT is diced with the desired pitch. This method requires each material stack layer to be separately prepared, aligned, and assembled, introducing multiple fabrication steps that increase complexity, processing time, and the likelihood of errors.

Optimal dicing outcomes depend on two dicing blade parameters. First, the size of the grits, composed of synthetic diamonds, influences the chipping mechanism during dicing[80]. Larger grit particles are more robustly anchored within the nickel binders, offering slower degradation rates. In contrast, smaller grit sizes are more advantageous for decreasing the amount of force on the transducer. This translates to finer dicing because it minimizes surface chipping and diminishes blade vibrations, resulting in a more precise cut. Second, the blade thickness determines the transducer kerf[80]. While a smaller thickness is preferable for achieving higher precision in transducer dicing, it increases the potential for blade vibration, which in extreme cases may lead to blade failure.

While utilizing smaller grit size and blade thickness is more advantageous for finer dicing, it introduces processing complexities. The conventional dicing strategy utilizes a full depth cut, wherein the entire transducer thickness (e.g., 1 mm in this study) is diced with a single pass (Supplementary Fig. 2)[84]. This strategy often induces substantial blade vibration due to the long contact edge between the blade and the transducer, which means that a large force is exerted on the blade, causing it to vibrate and potentially leading to blade breakage[80]. To minimize such vibration, a feed speed of 0.1 mm/s was typically utilized, extending the entire cutting process to ~15 h. Despite the low feed speed, this technique frequently caused thin blades to break. Thus, a larger blade thickness was required.

To address these challenges, in this study, we employed a super multipass dicing strategy, which dices each transducer element in incremental steps of 50 μm depth over a series of 20 rounds to achieve the full 1 mm depth (Supplementary Fig. 2). By reducing the depth of cut in each pass, this approach decreases the contact edge between the dicing blade and the transducer[80]. A shorter contact edge means that less force is exerted on the blade, which in turn reduces blade vibration and minimizes the risk of chipping the transducer (Supplementary Fig. 2). Therefore, the super multipass technique enables the use of thinner blades with smaller grit sizes, which generate transducers with narrower kerf without blade breakage. With such optimized processing, the transducer dicing can be performed at 10 mm/s, decreasing the total processing time to ~3 h (Supplementary Fig. 2).

The conventional method produced a 0.12 mm kerf, while the super multipass strategy produced transducers with a 0.05 mm kerf (Supplementary Figs 3 and 4). This preserved more active piezoelectric elements, improving the signal-to-noise ratio by 7.2 dB (Fig. 1b). Additionally, because the transducers were still fully diced, the crosstalk between individual channels was similar to that of the transducers diced with the conventional method (Supplementary Figs 3 and 4). The super multipass dicing method provides a one-step solution using a standard dicing

machine, eliminating the need for complex, multi-step fabrication while maintaining high dicing precision.

Supplementary Discussion 8: Acoustic lens

Ultrasound wave propagation is characterized by attenuation-induced energy loss (such as absorption, scattering, and reflection), resulting in a decreased signal-to-noise ratio as imaging depth increases[85].

The UPatch has an elevational length of 12 mm, providing a natural elevational focal depth of ~14 cm (Supplementary Fig. 7). Reducing the elevational length can reduce the natural focal depth, but this strategy compromises the acoustic energy output[85]. To ensure sufficient acoustic energy output while achieving optimal focal depth, we designed and fabricated an acoustic lens.

The acoustic lens focuses the ultrasound beam in the elevational dimension and concentrates energy into the lateral-axial imaging plane, thereby enhancing the signal-to-noise ratio[85]. A two-dimensional image in the lateral-axial plane assumes reconstruction from a thin slice but is actually from a thick slab due to the elevational beamwidth[85]. A large elevational beamwidth captures signals from both tissues and red blood cells for Doppler calculation, which compromises the signal-to-noise ratio[85,86]. The acoustic lens reduces signals from regions outside the area of interest and enhances the signal-to-noise ratio[87]. Therefore, in this study, we used an acoustic lens to focus the beam in the elevational direction, which reduces signals from regions outside the area of interest and enhances the signal-to-noise ratio[87].

The lens design is based on Snell's law:

$$\frac{\sin\theta_1}{\sin\theta_2} = \frac{v_2}{v_1} \tag{4}$$

where θ_1 is the incidence angle, θ_2 is the refraction angle, v_1 is the ultrasound wave velocity in the incident medium, and v_2 is the velocity in the refracted medium (Supplementary Fig. 5)[88].

Given a known speed of sound in the loading medium (such as biological tissues to be imaged), the acoustic lens modulates the ultrasound beam's shape depending on its speed of sound. Speed of sound is determined by the lens material. Rubbers are widely used due to their low acoustic impedance and simple fabrication process. The acoustic impedance of rubbers closely aligns with that of biological tissues (~1.68 MRayl), facilitating efficient ultrasound transmission into the tissue[89]. Additionally, rubber's low adhesion to metals enables facile molding from precisely machined metal molds[90]. Despite the relatively high acoustic attenuation of rubbers, the energy loss is minimal in imaging applications due to the small thickness of the lens.

The side of the lens tightly bonded to the transducer is flat, leading to an incidence angle of 90°, thereby eliminating refraction at the transducer/lens interface. The other side, which contacts the loading medium, is the interface where the refraction occurs. Given rubbers' lower speed of sound than water, simulations indicate that a convex lens with a constant radius effectively focuses the acoustic beam. The lens radius can be calculated as:

$$R_{lens} = F * |n - 1| \tag{5}$$

where F is the focal depth and R_{lens} is the lens radius. n is given by:

$$n = \frac{C_{medium}}{C_{lens}} \quad (6)$$

where C_{medium} is the speed of sound in the loading medium, and C_{lens} is the speed of sound in the lens[91].

We measured the acoustic properties of two silicone rubbers typically used in wearable ultrasound, Ecoflex and polydimethylsiloxane, using transmitters with broad bandwidths: P4-1 (ATL) for 1–4 MHz and L11-5 (Philips) for 5–10 MHz. The speeds of sound were 1007 m/s and 957 m/s, respectively (Supplementary Fig. 5). Despite Ecoflex exhibiting a slightly higher speed of sound, indicating a smaller lens radius and thicker lens, the difference between the two rubbers is negligible. Ecoflex has a lower attenuation coefficient than polydimethylsiloxane in the 1–10 MHz frequency range (Supplementary Fig. 5)[92-95]. Thus, we chose Ecoflex as the lens material. With a depth of interest of ~10 cm and an average speed of sound in tissue of 1540 m/s, we derived the lens radius to be 53 mm based on equations (5) and (6).

A three-dimensional model and its corresponding mold were designed in SolidWorks according to the transducer's dimensions (Supplementary Fig. 6). The designed mold was precisely machined using computer numerical control machining. Then, the lens was fabricated by casting and curing Ecoflex within the mold (Supplementary Fig. 6). The lens was then bonded to the transducer using Ecoflex to minimize the interfacial acoustic impedance mismatch.

We scanned the elevational-axial acoustic field to evaluate the lens' elevational beam focusing performance. Using the acoustic lens, the focal depth of the UPatch moved from 14 cm to 10 cm, which balanced the elevational resolution throughout the entire 20 cm depth of a typical womb (Fig. 1c and Supplementary Fig. 7). Additionally, the elevational beamwidth decreased from 6.0 mm to 4.5 mm at the focal point, improving the signal-to-noise ratio (Supplementary Fig. 7). While the acoustic lens enhances the signal amplitude at the focal region, it can also introduce minor attenuation due to the added lens material, which elongates the acoustic path and causes additional acoustic interfaces with impedance mismatch. These competing effects can offset each other. As a result, despite the more confined beam profile, the peak amplitude remains comparable at the focal point with and without the acoustic lens due to the attenuation introduced by the lens (Supplementary Fig. 7).

Beam steering controls lateral positioning in the XZ plane, while the acoustic lens enhances elevational beam focusing in the YZ plane to improve signal quality. The acoustic lens focusing remains beneficial even when the beam is steered off-axis in the XZ plane. When steered from -7.5° to 7.5°, the angle range of the duplex imaging algorithm used in this work, the elevational beamwidth is consistent at ~4.5 mm and ~6.0 mm for with and without the acoustic lens, respectively (Supplementary Fig. 8). The improved beam confinement leads to higher contrast and better vessel visualization, ensuring more reliable blood flow measurements.

Supplementary Discussion 9: Soft Faraday cage

The UPatch is vulnerable to electromagnetic interference due to its operational frequency within the radiofrequency spectrum. These interferences primarily originate from two sources: external electronic devices and contact with the human body.

External electronic devices (e.g., smartphones, Wi-Fi routers, and medical equipment) emit electromagnetic signals to the surrounding environment, coupling to the UPatch[96]. The in-band electromagnetic noise at 2.5 MHz is unlikely to couple into the UPatch through the air due to its long wavelength (120 m). However, these electronic devices also emit electromagnetic signals at frequencies different from ultrasound signals, which can still contribute to in-band interference through several mechanisms. First, external electromagnetic signals can couple with the UPatch due to its physical dimensions being comparable to the wavelength. For example, Wi-Fi signals operate at 2.4 GHz, which corresponds to a wavelength of ~12.5 cm. This wavelength is comparable to the size and geometry of the UPatch, allowing electromagnetic interference to occur. Second, intermodulation of various external electromagnetic signals can cause down conversion to in-band interference[97]. Multiple electromagnetic signals can undergo nonlinear mixing within the backend system, generating new frequency components through intermodulation[97]. These newly generated signals can span a broad range of frequencies, including those overlapping with ultrasound signals, potentially degrading signal quality and affecting system performance. Third, ground loops can allow external electromagnetic signals to couple into the system, generating low-frequency interference[98]. In a single-ended configuration, ground loops can behave like coils, introducing parasitic inductance that enables inductive coupling with external electromagnetic fields, further exacerbating interference[99].

Direct contact with the human body transmits noise to the UPatch due to its biopotentials and antenna effects. Biopotentials (such as electrocardiogram and electromyogram) generated from the human body can also be modulated by external electromagnetic signals, resulting in noises near the ultrasound frequency band[100,101]. The antenna effects are a phenomenon where the human body inadvertently acts as an antenna that couples electromagnetic waves in the air[102].

Ultrasound signals collected from deep tissues are intrinsically weak[85]; thus, these electromagnetic interferences can degrade signal and image quality. Consequently, devising effective screening strategies is imperative for the UPatch to image the fetus and acquire high-fidelity spectral Doppler signals from fetal vessels.

Various methods can reduce electromagnetic noise. First, analog and digital filters (such as low-pass, high-pass, and band-pass filters) are commonly used to attenuate noise outside the frequency band of interest[103]. The backend system (Vantage 256, Verasonics) incorporates these filters, effectively eliminating outside-band noise (such as biopotentials) which are typically below 10 kHz[104]. However, due to the wideband nature of electromagnetic noise in the air, filters alone are insufficient for lowering the noise floor of in-band signals. Second, specialized circuits can reduce thermal noise (caused by irregular movement of electrons inside electronic components due to ambient temperature) coupled to ultrasound signals during the filtering, amplification, and digitization processes[103]. However, this method increases the footprint, making the UPatch less wearable. Third, ground shielding the entire surroundings of the UPatch can reduce the noise floor by directly blocking the noise.

Electromagnetic interference is ubiquitous and omnidirectional, and the most effective shielding approach is a Faraday cage to prevent unwanted coupling from various sources[103]. These three-dimensional shields redistribute the charges across the cage surface, thereby neutralizing both static and dynamic electrical fields[103]. Metals are often used to construct the cage due to their high conductivity for the charges to move freely and quickly for effective redistribution[103]. A key design guideline to develop a soft Faraday cage is based on fundamental wave propagation theory. The

effectiveness of a Faraday cage in blocking electromagnetic signals depends on the spacing between stretchable traces, which is governed by:

$$L < \frac{c}{10f\sqrt{\epsilon_r}} \quad (7)$$

where L is the spacing between adjacent net traces, c is the speed of light, f is the maximum frequency of the blocked signals, and ϵ_r is the relative permittivity[105].

We designed a soft Faraday cage to shield the UPatch from electromagnetic noise. The previous design incorporated two isolated copper electrode layers[73], which provided two-dimensional shielding but could not block electromagnetic interference from entering through the edges of the two electrodes[75]. In contrast, the UPatch was designed with a soft Faraday cage that was made of laser-ablated polyimide/copper laminates and vertical interconnect accesses. The soft Faraday cage connects the edges of the two electrodes using stitching vertical interconnect accesses. The transducers were placed between the mesh ground electrode and electromagnetic shielding electrode (Supplementary Fig. 9). Most of the body noise coupled to the UPatch will be shunted to the mesh ground electrode close to the skin. Various electromagnetic frequency signals can undergo intermodulation, generating spurious low-frequency signals that may overlap with the ultrasound frequency band, leading to interference[98]. The added electromagnetic shielding layer to the top of the signal electrodes can potentially shield noise from the air. The closely patterned electrical connections prevent electromagnetic interference from entering the UPatch from the edges, providing three-dimensional shielding in all directions (Supplementary Fig. 10). The maximum spacing between adjacent net traces was 0.9 mm, which can effectively block electromagnetic waves up to 20 GHz, covering all frequency bands used in common electronic devices[103].

We analyzed the power spectral density, which quantitatively measures the effectiveness of the soft Faraday cage[106]. In radiofrequency electronics (such as cellular transceivers, Bluetooth, and Wi-Fi), dBm/Hz is the standard unit in power spectral density plot because the signal power in these systems is typically in the milliwatt range[107]. However, from a general signal processing perspective, dB/Hz is a more appropriate unit for describing spectral density in relative terms[108]. The baseline noise level was also referenced to 1 W, ensuring consistency in the relative power representation.

We evaluated baseline noise levels of the backend system, UPatch without the soft Faraday cage, and UPatch with the soft Faraday cage. The baseline noise levels of the backend system were measured in two conditions: input termination at 50 Ω and open circuit. The input termination at 50 Ω matches the system's impedance and minimizes electromagnetic interference, while the open circuit captures the system's overall susceptibility to environmental noise. The baseline noise levels under 50 Ω input termination and open circuit were negligibly similar (Fig. 1d). This observation indicates that the noise is primarily contributed by the system itself rather than by external sources. Therefore, the measured baseline noise level can be considered a reliable representation of the system's intrinsic noise, regardless of the termination condition[109]. The noise level of the UPatch with the soft Faraday cage closely mirrored the baseline noise level of the backend system, which is mainly composed of thermal noise (Fig. 1d). In the absence of the soft Faraday cage, the noise floor increased by 11.7 dB/Hz at 2.5 MHz, the center frequency of the UPatch.

We also performed measurements under both phantom and in vivo conditions (Supplementary Figs 11–13). Noise levels were characterized using the probability density function, which captures the distribution of instantaneous signal amplitudes over time[110]. Broader distributions indicate greater variability due to environmental or physiological noise, making it suitable for statistically quantifying noise[110]. Phantom measurements revealed the UPatch's susceptibility to ambient electromagnetic noise, while in vivo measurements demonstrated additional susceptibility to physiological noise. The soft Faraday cage effectively attenuated environmental interference from nearby electronic devices (Supplementary Fig. 11) and reduced noise contributions from both environmental and physiological sources, resulting in improved signal fidelity (Supplementary Fig. 12). As a result, duplex imaging with the soft Faraday cage showed enhanced image quality and reduced noise in both phantom and in vivo settings (Supplementary Fig. 13).

The soft Faraday cage provided shielding against electromagnetic noise while conforming to the skin. Thus, this shielding strategy can be effectively adapted to enhance electromagnetic compatibility and reduce noise in soft electronics.

Supplementary Discussion 10: Umbilical cord phantom

Commercially available fetal phantoms primarily support structural B-mode imaging[111]. To the best of our knowledge, no commercially available phantoms accurately replicate the complex flow dynamics and anatomical variability of the umbilical cord and fetus.

To further validate the UPatch's ability to detect the umbilical cord in varying orientations, we developed an umbilical cord phantom that replicated the anatomical complexity of the intertwined arteries and veins in the cord. The phantom design incorporated two helically wrapped arteries (diameter ~3 mm) around a vein (diameter ~6 mm) to simulate the intertwined vascular geometry observed in vivo (Supplementary Fig. 18)[112]. We used deionized water as the surrounding medium, reflecting the high water content (~98%) of amniotic fluid to replicate the in vivo acoustic environment (Supplementary Fig. 18)[113].

Commercially available silicone tubing (Metaland) does not accurately mimic the acoustic impedance of soft tissue (~1.65 MRayl)[85]. Its substantially lower impedance (~1.07 MRayl) results in pronounced mismatch at fluid–structure interfaces, producing strong reflection during insonation across multiple vascular boundaries of the umbilical cord (Supplementary Fig. 18)[114]. To address this limitation, we selected Elastic 50A resin (Formlabs), which has acoustic impedance (~1.66 MRayl) that closely aligns with the acoustic impedance of soft tissue (Supplementary Fig. 18)[115,116]. Using the Elastic 50A resin, high-resolution stereolithographic printing enabled precise and repeatable fabrication of the complex intertwined geometry of the umbilical cord (Supplementary Fig. 19).

We conducted duplex imaging of the umbilical cord phantom across a range of vessel orientations to mimic real-world dynamics caused by fetal and maternal movements (Extended Data Fig. 2). These experiments demonstrated that the UPatch could reliably acquire the umbilical vessel flow signals despite dynamic anatomical orientation changes in the in-utero environment.

Supplementary Discussion 11: Acoustic exposure safety parameters

The UPatch was activated at 30 V with a pulse repetition frequency (PRF) of 3000 Hz, which were the highest values used in this study.

Using power P from a radiation force balance or a hydrophone (Supplementary Figs 23 and 24), spatial average temporal average intensity (I_{SATA}) can be calculated by:

$$I_{SATA} = \frac{P}{A} \tag{8}$$

where A is the area of the aperture (306.6 mm^2 in this study)[117]. This equation is valid for operation with an unfocused wave, making it applicable to brightness and color-flow modes in this study[117].

After waveform deconvolution using the data from calibrating the hydrophone sensitivity (Supplementary Fig. 23), the hydrophone measurements calculated the derated pressure as:

$$p_\alpha(z) = p(z)10^{\left(-\frac{\alpha z f_{awf}}{20dB}\right)} \tag{9}$$

where α is the attenuation coefficient (conservatively assumed to be 0.3 $dB \cdot cm^{-1} \cdot MHz^{-1}$)[117], z is the distance from the UPatch along the beam axis to the plane containing the point of interest, and f_{awf} is the acoustic working frequency. Then, the derated pulse pressure squared integral can be calculated as:

$$ppsi_\alpha(z) = \int p_\alpha^2(z,t)dt \tag{10}$$

where measurements are based on the maximum value of $ppsi_\alpha(z)$ between the break-point depth z_{bp} and focal depth[118]. And z_{bp} is defined as:

$$z_{bp} = 1.5 \times D_{eq} \tag{11}$$

where D_{eq} is the equivalent aperture diameter and is defined as:

$$D_{eq} = \sqrt{\frac{4}{\pi}A} \tag{12}$$

Thus, z_{bp} is estimated to be 29.64 mm.

The derated pulse intensity integral is calculated by:

$$pii_\alpha(z) = \frac{1}{\rho c} ppsi_\alpha(z) \tag{13}$$

where ρ is the density of water (997 $kg \cdot m^{-3}$) and c is the speed of sound in water (1480 $m \cdot s^{-1}$)[85]. Therefore, the derated spatial peak pulse average intensity is defined as:

$$I_{SPPA,\alpha}(z) = \frac{1}{t_d(z)} pii_\alpha(z) \tag{14}$$

where $t_d(z)$ is the pulse duration, which is 1.25 multiplied by the interval between the time when the time integral of the square of the instantaneous acoustic pressure reaches 10% and 90% of its final value[119,120].

The derated spatial peak temporal average intensity can be calculated as:

$$I_{SPTA,\alpha}(z) = PRF\ pii_{\alpha}(z) \tag{15}$$

The mechanical index (MI) describes the relative probability of eliciting micromechanical bio-effects, such as those associated with cavitation-induced damage[117,121], and is defined as:

$$MI = \frac{p_{r,\alpha}(z_{MI})}{\sqrt{f_{awf}}} \tag{16}$$

where $p_{r,\alpha}$ represents the derated peak rarefactional pressure in MPa, measured along the beam axis at depth z_{MI} where $ppsi_{\alpha}(z)$ is maximum, and f_{awf} is in MHz.

The thermal index (TI) is an index of the estimated temperature rise in tissues and is defined as:

$$TI = \frac{Acoustic\ power\ output}{Acoustic\ power\ ouput\ needed\ to\ produce\ 1\ ℃\ rise} \tag{17}$$

TI estimates temperature rise, postulating that the attenuation within tissue is homogeneous with an attenuation coefficient of 0.3 dB·cm^{-1}MHz^{-1}, which is lower than the attenuation coefficient of most soft tissues[122]. Depending on the tissues in the scanning plane, *TI* is divided into three types: soft tissue TI (TIS), bone TI (TIB), and cranium TI (TIC)[118,122].

First, the TIS is the maximum of the calculated values from equations (18) and (19):

$$TIS_{as,ns} = \frac{P_{1\times1}\cdot f_{awf}}{210\ mW\ MHz} \tag{18}$$

where $TIS_{as,ns}$ is the TIS at surface (non-scanning), $P_{1\times1}$ is the bounded–square output power over a 1 cm^2 area.

$$TIS_{bs,ns} = \min\left[\frac{P_{\alpha}(z_{s,ns})\times f_{awf}}{210\ mWMHz}, \frac{I_{SPTA,\alpha}(z_{s,ns})\times f_{awf}}{210\ mWcm^{-2}MHz}\right] \tag{19}$$

where $TIS_{bs,ns}$ is the TIS below surface (non-scanning), $P_{\alpha}(z_{s,ns})$ is the derated output power at depth $z_{s,ns}$, and $I_{SPTA,\alpha}(z_{s,ns})$ is the derated spatial peak temporal average intensity at depth $z_{s,ns}$.

The derated power can be calculated as:

$$P_{\alpha}(z_{s,ns}) = P(z_{s,ns})10^{\left(-\frac{\alpha z f_{awf}}{10\ dB}\right)} \tag{20}$$

where depth $z_{s,ns}$ is defined as:

$$z_{s,ns} = depth\ of\ max\{min[I_{spta,\alpha}(z)\times 1\ cm^2, P_{\alpha}(z)]\} \tag{21}$$

The TIB is calculated as:

$$TIB = \min\left[\frac{\sqrt{P_{\alpha}(z_{b,ns})\times I_{SPTA,\alpha}(z_{b,ns})}}{50\ mWcm^{-1}}, \frac{P_{\alpha}(z_{b,ns})}{4.4\ mW}\right] \tag{22}$$

where depth $z_{b,ns}$ is defined as:

$$z_{b,ns} = depth\ of\ max\{P_\alpha(z) \times I_{SPTA,\alpha}(z)\} \quad (23)$$

The TIC can be calculated by:

$$TIC = \frac{P/D_{eq}}{40\ mW cm^{-1}} \quad (24)$$

For duplex imaging, each TI based on B-mode and color Doppler modes was calculated separately and summed. The TI for Spectral Doppler mode was calculated for each focusing depth, and the highest value was reported.

Supplementary Discussion 12: Thermal safety of the UPatch

Safety guidelines for obstetric ultrasound are provided by the American Institute of Ultrasound in Medicine (AIUM)[123], British Medical Ultrasound Society (BMUS)[124], and Food and Drug Administration (FDA)[117]. The UPatch is an ultrasound device that may expose the mother and fetus to various safety hazards. Therefore, the device must be characterized accurately to meet the recommendations of the regulatory bodies.

The FDA has different recommendations for Track 1 and Track 3 devices (Track 2 is not available due to historical reasons)[117]. The UPatch was characterized to meet the recommendations for both Track 1 and Track 3 devices to ensure rigorous thermal safety evaluation.

The FDA Track 1 recommendations provide maximum values for the I_{SATA}, $I_{SPPA,\alpha}$, $I_{SPTA,\alpha}$, and MI for fetal applications. These recommended maximum values are I_{SATA} < 20 mW/cm^2, $I_{SPPA,\alpha}$ < 190 W/cm^2, $I_{SPTA,\alpha}$ < 94 mW/cm^2, and MI < 1.9 (ref. [117]). The I_{SATA} of the UPatch (0.72 mW/cm^2) is substantially lower than the recommended maximum value and the Corometrics 250CX Series (GE Healthcare)[18] fetal heart rate monitor, which is reported to be < 5 mW/cm^2 (Supplementary Fig. 24). For $I_{SPPA,\alpha}$, the measured values from the hydrophone were 4.51 W/cm^2 for duplex imaging and 16.42 W/cm^2 for spectral Doppler. Additionally, the $I_{SPTA,\alpha}$ values were 19.86 mW/cm^2 for duplex imaging and 58.5 mW/cm^2 for spectral Doppler. The MI was 0.20 for duplex imaging and 0.37 for spectral Doppler (Fig. 1f). The UPatch operates well below the FDA Track 1 recommended maximum values.

The FDA Track 3 recommendations require device manufacturers to display MI and TI values according to the Output Display Standard in IEC 60601-2-37 (ref. [117,125]). MI and TI values are displayed in real time to ensure that clinicians adhere to the "as low as reasonably achievable" principle, minimizing patient exposure to ultrasound energy – although evidence suggests these are rarely checked in practice[126]. While there is no specific recommendation from the FDA[117] for the maximum TI, an explanation is requested for TI over six. The AIUM[123] and BMUS[124] set recommendations for maximum dwell time based on TI values. According to these guidelines, the recommended dwell time for obstetric scanning is set as: no dwell time limit for $TI \leq 0.7$, maximum dwell time < 60 min for $0.7 < TI \leq 1.0$, maximum dwell time < 30 min for $1.0 < TI \leq 1.5$, maximum dwell time < 15 min for $1.5 < TI \leq 2.0$, maximum dwell time < 4 min for $2.0 < TI \leq 2.5$, maximum dwell time < 1 min for $2.5 < TI \leq 3.0$, and scanning with $TI > 3.0$ is not recommended for obstetric uses.

The maximum increase in the surface temperature of the UPatch in air was measured to be 0.53 °C (Supplementary Fig. 22). The device maintained a stable surface temperature well below safety thresholds, supporting its viability for extended clinical use. Using measurements from a radiation force balance or a hydrophone, we calculated the highest TI values for duplex imaging and spectral Doppler. For duplex imaging, the TIS was 0.02, TIB was 0.05, and TIC was 0.05 (Fig. 1f). For spectral Doppler, the TIS was 0.10, TIB was 0.38, and TIC was 0.17 (Fig. 1f). All of these values were < 0.7, suggesting no dwell time limit for the UPatch according to AIUM and BMUS recommendations[123,124]. For obstetric imaging, the TIS is most relevant thermal index up to 10 weeks from the last menstrual period or a crown-rump length of about 33–34 mm, and TIB is the most relevant thereafter[123].

In summary, the UPatch adheres to recommendations by the FDA Track 1 and Track 3 (ref. [117]), AIUM[123], and BMUS[124]. Additionally, we strictly followed the as low as reasonably achievable principle to minimize ultrasound wave exposure to the fetus and the mother[117].

Supplementary Discussion 13: Beamforming of the UPatch on the abdomen

Beamforming is traditionally performed using a fixed array configuration. Thus, the UPatch image reconstruction assumes that the 64 channels are on a static surface. However, the curvature of the pregnant woman's abdomen differs from person to person and varies at different locations of the abdomen, introducing phase aberration to the beamforming of the UPatch[127]. This curvature for the third trimester typically has a radius ranging from 14 cm to 18 cm[128,129].

To mitigate this problem, phase aberration was corrected for a curvature radius of 16 cm, accommodating the typical abdominal geometry (Supplementary Fig. 25). The aberration correction was applied in both transmission and reception, which adjusted the time delay profile of the ultrasound waves to ensure accurate image reconstruction. The delay-and-sum beamformer used the same curvature profile for consistent aberration correction during reception. With this phase aberration correction, the UPatch could provide high-quality images even with varying maternal abdominal curvatures (Extended Data Fig. 3 and Supplementary Figs 27 and 28). Additionally, while the curved shape of the maternal abdomen focuses the ultrasound beam, we compensated for the bending radius by beamforming, so unwanted focusing was largely alleviated.

Supplementary Discussion 14: Vessel tracking algorithms

Traditional methods require clinicians to manually maneuver the handheld clinical ultrasound device and select the appropriate sample gate on the duplex image for spectral Doppler analysis of fetal vessels (Supplementary Fig. 31)[53]. This approach can only provide intermittent measurements and faces challenges during dynamic physiological changes (e.g., fetal and maternal movements, respiratory processes, and uterine contractions)[48,51]. These changes can alter the locations of fetal vessels and sample gates[130,131]. For example, the umbilical cord can rotate and orient in various directions[53]. Such movements can displace the sample gate initially placed on the umbilical artery to other tissues (Fig 3c). Thus, strategies are needed to adaptively adjust the sample gate to the moving vessel.

Efforts to address this challenge remain in early stages of development. One approach involved developing a vessel tracking algorithm that detects the vessel wall using A-mode[132]. However, this algorithm cannot differentiate between various vessel types (such as the umbilical arteries and vein) or vessel orientations (such as the transverse and longitudinal planes). In a separate study, computer vision was used to estimate the ultrasound Doppler angle for automatic angle correction. Yet this method still requires a clinician to initially identify the vessel of interest[133]. Furthermore, a machine learning model applied to M-mode images successfully detected the vessel diameter but was limited to the transverse view of the vessel[134].

Image segmentation offers a promising alternative for vessel detection because different vessel types and orientations provide distinct features. Thus, commercial devices have recently implemented image segmentation-based strategies. The Arietta 65 (Fujifilm) claims to automate sample gate selection for mitral valve in the apical four-chamber view[135] and received FDA clearance in November 2020 (ref. [136]), but the automated sample gate feature[135] was not included in its regulatory submission. The Acuson Origin (Siemens) received FDA clearance in October 2023 (ref. [137]). This artificial intelligence system places sample gates for 12 cardiac views, supporting up to 4 anatomical locations with a total of 42 sample gate placements[137,138]. The automated sample gate feature in Kosmos (cart-based ultrasound system; EchoNous) and Kosmos Mobile (wireless ultrasound system; EchoNous) was introduced in September 2024 (ref. [139-141]) but has not received FDA clearance[142]. An artificial intelligence system was proposed to provide sample gate selection for apical four-chamber view, specifically for the left ventricular outflow tract, mitral valve, pulmonary valve, and tricuspid valve[140].

These developments reflect exciting progress in echocardiography. However, there remains no commercially available solution for automated sample gate selection in fetal ultrasound. Unlike cardiac imaging, which benefits from anatomical landmarks that are changing periodically, fetal imaging must contend with continuously shifting anatomy due to fetal and maternal movements. The intertwined vascular geometry of the umbilical cord, comprising two arteries helically wrapped around a central vein, results in dynamic variation in depth, orientation, and position (Extended Data Fig. 2). Additionally, these existing commercial probes must be held by humans or robotic arms, making them impractical for continuous monitoring.

The image segmentation in this study utilizes color Doppler signals in the duplex image. This method effectively differentiates the umbilical artery from the umbilical vein by identifying differences in the flow direction and pulsatility (Fig. 3a). It enables dynamic tracking and adaptive sample gate selection in response to changing vessel geometry and motion, representing a substantial advance over existing cardiac-focused approaches and addressing an unmet need in continuous fetal blood flow monitoring. Additionally, we optimized the computational load by reducing the number of frames analyzed by the tracking algorithm. The UPatch is self-adhering and integrates autonomous vessel tracking specifically designed for fetal ultrasound, enabling prolonged monitoring without the need for a sonographer or a robotic arm. This method facilitates autonomous monitoring during physiological changes, thereby enhancing the reliability and effectiveness for fetal monitoring.

Supplementary Discussion 15: Working principles of vessel tracking algorithm

The tracking algorithm leverages pulsatility characteristics encoded in the color Doppler signal to quantitatively differentiate the vessels.

Each pixel in the frame is assigned a color intensity value ranging from 1 to 128 (Extended Data Fig. 6). Values from 1 to 64 represent flow away from the UPatch using cool hues, where lower color intensity values correspond to cyan (higher velocity), and higher values correspond to dark blue (lower velocity). Values from 65 to 128 represent flow toward the UPatch using warm hues, where lower color intensity values correspond to dark red (lower velocity), and higher values correspond to yellow (higher velocity).

Based on this numerically assigned pixel color intensity, the inter-frame color difference compares the color intensity of each pixel among consecutive frames. A high inter-frame color difference reflects greater temporal variation in flow velocity, characteristic of arterial pulsatility. In contrast, low variation corresponds to steadier flow typically observed in veins. Thus, the algorithm evaluated the inter-frame color difference to obtain pixels designated for the umbilical artery or vein. We implemented a threshold for the inter-frame color difference across each vessel region. Regions exceeding this threshold are classified as arteries, while those below are identified as veins.

Binary segmentation maps were generated, followed by morphological opening and closing to remove granular noise and smooth region boundaries. Opening removes small, isolated artifacts, while closing fills gaps and connects nearby regions[143,144]. These two steps improve segmentation quality by preserving large, coherent vessel structure. Among the identified regions, the one with the largest pixel area is designated as the primary region because larger regions offer more stable signals, are less affected by noise, and align with the selection norm typically used by sonographers in clinical practice (Supplementary Fig. 33). This approach effectively suppresses sparse, spatially dispersed noise while retaining the dominant vascular structure. Finally, we computed the geometric centroid of the selected primary region to define the sample gate location (Supplementary Fig. 34). The centroid, calculated as the mean x and y coordinates of all pixels within the segmented region, ensures a spatially robust placement of the sample gate.

Supplementary Discussion 16: Umbilical vein tracking

Most clinical applications of fetal Doppler focus on the umbilical artery[48,51,67]. Therefore, this study optimized the tracking algorithm specifically for the detection of the umbilical artery (Fig. 3a).

Nevertheless, this tracking algorithm can also be adapted to detect the umbilical vein. The umbilical vein transports oxygenated blood from the placenta to the fetus, ensuring it receives essential oxygen and nutrients for its growth and development (Supplementary Discussion 5)[145]. Thus, monitoring the umbilical vein is likely to provide direct insights into fetal oxygenation.

While the umbilical arteries and vein are intertwined, their color Doppler signals are typically opposite due to their opposite flow directions (Extended Data Fig. 3). Thus, if the color Doppler signal of the arteries is red (indicating flow towards the UPatch), the signal of the vein will be blue (indicating flow away from the UPatch), or vice versa. Additionally, the umbilical vein is typically

non-pulsatile because the placenta acts as a blood reservoir (Supplementary Discussion 5)[48]. Therefore, when the tracking algorithm calculates the variance in color Doppler pixel intensity over several consecutive frames, it can identify the umbilical vein as the non-pulsatile region (Fig. 3a). A similar segmentation method can then be applied to identify the primary region and its spatial centroid of the umbilical vein for sample gating (Fig. 3a). This adaptation allows the tracking algorithm to be utilized for continuous umbilical vein tracking and blood flow monitoring.

Supplementary Discussion 17: Response time during image segmentation

Due to limitations in the computational power of the backend system, the autonomous tracking algorithm cannot simultaneously perform duplex imaging, tracking, and spectral Doppler recording. Thus, the algorithm sequentially reconstructs a duplex image, tracks the umbilical artery to obtain a sample gate, and records spectral Doppler signals (Fig. 3a). In such cases, fetal movement may affect the performance of the image segmentation algorithm. Fast movements can particularly cause the initially registered sample gate to shift away from the target vessel before the next round of image segmentation begins.

To address this issue, we integrated a standard deviation threshold at the end of the image segmentation sequence. This ensures that when the algorithm detects fast cord movements, it does not provide a sample gate until the movement is slowed down. The algorithm computes the x and y coordinates of tracked points across consecutive frames. A threshold based on the standard deviation of these coordinate changes was applied to identify and exclude tracked points with abrupt displacements, which may occur during rapid fetal movements. This approach reduces the likelihood of erroneous sampling and ensures reliable tracking performance during continuous monitoring of dynamic conditions. Consequently, sample gates were applied only to vessels that have no or slow movements (Supplementary Video 1).

A typical uninterrupted period of observations is ~ 5 s. Fetal and maternal movements are the dominant factors affecting continuity, and we did not observe dependence on gestational age or pathology grouping. A minimum uninterrupted duration of ~2 s (corresponding to >5 fetal cardiac cycles) is required to localize and stably track the umbilical artery in duplex imaging for spectral Doppler acquisition. During continuous monitoring sessions, duplex mode vessel tracking occurred in intermittent segments of variable duration (typically several seconds), reflecting ongoing fetal and maternal motion, while still allowing recurrent spectral Doppler acquisition whenever motion was sufficiently slow (Supplementary Video 1).

Despite this improvement, the method still has limitations. First, the autonomous tracking algorithm cannot detect movements after spectral Doppler signals have been acquired. Second, waiting until the vessel stops moving quickly to apply a sample gate leads to variability in the response time. Third, the algorithm cannot acquire spectral Doppler during vessel movement.

To mitigate these issues, three solutions are proposed. First, an additional algorithm can be developed to identify signal loss in the spectral Doppler based on its luminosity[134]. This would detect fast movements, enabling the algorithm to identify when the signal is lost and switch back to duplex imaging for further tracking. Currently, we minimized this issue by acquiring 3 s of spectral Doppler prior to switching back to duplex imaging (Fig. 3c and Supplementary Video 1).

This approach ensures that spectral Doppler signals are acquired for a short duration, minimizing the impact of movement.

Second, image segmentation-based machine learning models can directly generate a sample gate from a single image[146]. The current tracking algorithm requires multiple frames to calculate the sample gate, resulting in variability in response time (Supplementary Video 1). To solve this problem, machine learning models, trained on annotated datasets, allow individual images to generate a sample gate. This approach minimizes variability in response time and tracks the target vessel during movement.

Third, integrating a more advanced graphics processing unit in the backend system can substantially enhance the beamforming framework[147]. Recent advances in graphics processing units enable a much larger number of parallel executing cores, facilitating faster computation and allowing simultaneous duplex imaging and spectral Doppler acquisition[148]. This approach allows spectral Doppler acquisition even during movement, thereby improving overall efficiency and response time.

Supplementary Discussion 18: Evaluation of the autonomous tracking algorithm

The vessel tracking algorithm is based on conventional image processing techniques, which do not require cross-validation. Machine learning or artificial intelligence-based models require a labeled training dataset to learn features and a separate testing dataset to evaluate performance. Our approach applies deterministic operations directly to the input data and does not involve model training. There is no differentiation between training and testing datasets. Therefore, cross-validation procedures, which are essential for evaluating generalizability in trained models, are not applicable in this context.

The autonomous tracking algorithm in this study was evaluated using three methods to test its accuracy and reliability across different scenarios. In fetal ultrasound, manual sample gating by experienced sonographers remains the clinical gold standard for assessing signal quality and tracking accuracy[149]. The evaluation method we employed aligns with established clinical standards and other studies in the literature[137,150]. Additionally, in contrast to these cited studies[137,150], which employed only a single validation method, our approach was comprehensively validated using three independent methods. In the FDA-cleared Acuson Origin (Siemens) system, the autonomous sample gate placement underwent validation using only 168 images as part of its FDA clearance process[137]. The validation accuracy evaluated by three sonographers was defined as "no adjustment or only minor adjustment needs to be made by the user"[137]. In the peer-reviewed study, accuracy of a deep learning-based sample gating was validated using a method similar to our first validation method[150]. To the best of our knowledge, these reported works were the only studies that demonstrate tracking algorithm validation for ultrasound imaging.

The first two methods were based on pre-collected images, allowing direct comparison against sonographers. In the first method, a double-blind setup was used where both the tracking algorithm and the sonographer selected optimal tracking points in the same images collected using the UPatch. The differences between these points were then measured in axial and lateral dimensions, providing a quantifiable metric of the algorithm's accuracy.

In the second method, three sonographers visually assessed the accuracy of the tracking algorithm. They categorized the tracked points into three levels: "Optimal" for points suitable for acquiring high-quality spectral Doppler signals, "Sub-optimal" for points that might yield low quality signals, and "Non-diagnostic" for points incapable of providing usable Doppler signals. However, these evaluations did not account for the actual spectral Doppler signals that could be acquired from the tracked points.

Thus, a third method was employed to quantify the differences between the systolic-to-diastolic ratios of the actual spectral Doppler signals collected from the points of the autonomous tracking algorithm and those selected by a sonographer, providing a more practical assessment of the algorithm's performance.

This three-step evaluation process provided a robust framework for optimizing the tracking algorithm. By incorporating offline sonographer comparisons of the sample gates and practical online assessments of spectral Doppler accuracy, the study ensured the algorithm was finely tuned for clinical uses.

We also performed a retrospective comparison with two widely used deep learning models: U-Net[151] and ConvLSTM[152] (Supplementary Fig. 35). Both deep learning models were trained and evaluated using a dataset containing 1,000 duplex images acquired with the UPatch. The dataset was split into independent training and validation sets, and training was continued until convergence of the validation loss to ensure optimal model performance without overfitting.

Tracking accuracy was assessed by three independent sonographers. The tracking algorithm in this work achieved a tracking accuracy of 93.1%, the U-Net model achieved 92.2%, and the ConvLSTM model achieved 96.7%. Computational efficiency was evaluated by measuring per-frame processing time on the backend system (Vantage 256, Verasonics). The tracking algorithm in this work required 0.0698 ± 0.0083 s per frame, the U-Net model required 0.719 ± 0.0436 s, and the ConvLSTM model required 0.7439 ± 0.0535 s. While the accuracy of the three methods is similar, the tracking algorithm developed in this work achieves more than a tenfold reduction in processing time, enabling real-time image processing.

Supplementary Discussion 19: Ultrasound sample gate

Accurately delineating the sample gate within the duplex image is crucial[153]. Ideally, the sample gate should cover the entire vessel diameter to ensure accurate measurement of the complete hemodynamic profile[48]. However, clinicians commonly position a 2 mm sample gate at the vessel's center to record the spectral Doppler signals (Supplementary Fig. 31)[50]. There are two practical reasons.

First, the 2 mm size ensures that the vessel does not shift away from the sample gate during measurements[48]. The umbilical arteries are closely intertwined with the umbilical vein, necessitating a clear distinction between them. This sample gate size is typically smaller than most fetal vessels in the third trimester, allowing assessment of the target vessel even during small vessel movements[53].

Second, Doppler indices derived from the vessel's center are negligibly different from those derived from the entire vessel diameter. Blood flow within vessels typically exhibits a parabolic

profile (laminar flow), with red blood cells traveling at varying velocities[85]. This velocity gradient is characterized by the highest velocities at the vessel's center and the lowest velocities near the vascular wall due to viscous drag[85]. Doppler indices, such as the systolic-to-diastolic ratio, pulsatility index, and resistance index, are derived from the peak systolic velocity, end diastolic velocity, and mean flow velocity extracted from the spectral waveform envelope[48,51]. The peak systolic velocity, end diastolic velocity, and mean flow velocity extracted from the spectral waveform envelope of the entire vessel diameter are negligibly different from those extracted from the spectral waveform envelope of the vessel center. Thus, these Doppler indices are not affected much when the sample gate is positioned at the vessel's center.

Because of such practical considerations, this study utilized the standard sample gate size of 2 mm.

Supplementary Discussion 20: Continuous monitoring of the umbilical artery

Doppler indices vary along the umbilical artery, with blood flow resistance highest at the fetal abdominal cord insertion, intermediate at the free-loop cord, and lowest at the placental cord insertion[49]. Thus, it is important to sample spectral Doppler consistently at the same site of the umbilical artery throughout the duration of continuous monitoring[60].

Continuous monitoring of the umbilical cord at the fetal abdominal insertion site presents practical difficulties, particularly in advanced gestation[49]. First, the fetus can move or rotate in the uterus, causing the umbilical cord at the fetal abdominal insertion site to move out of the sonographic window. Second, the site may be obscured by fetal lower limbs or oligohydramnios, a condition characterized by low amniotic fluid levels that limit the acoustic window[49].

Sampling a fixed position of the free-loop cord also has disadvantages. First, the umbilical cord can extend an average length of 64 cm, making it challenging to sample consistently at the same position during continuous monitoring[154]. The umbilical cord is constantly displaced as it floats inside the amniotic fluid under various environmental changes, including fetal movement, fetal breathing, maternal movement, maternal breathing, and uterine contractions[51].

The umbilical cord at the placental cord insertion is also moving, but the placenta is fixed on the maternal uterine wall[48,51]. Although the umbilical cord itself moves, its position at the placental cord insertion remains relatively identifiable by using the placenta as a stable anatomical reference (Extended Data Fig. 5)[50]. As a result, the UPatch was positioned to insonate the umbilical cord at the placental cord insertion. In our experiments, after identifying the optimal location and orientation using a clinical device, we laminated the UPatch. Throughout various maternal and fetal movements, this consistent anatomical reference enables reliable imaging at the placental cord insertion, ensuring continuous monitoring without relocating the UPatch.

We conducted a study in which the UPatch was tested across a range of common maternal postures: lying supine, lying on the right side, lying on the left side, and sitting upright (Extended Data Fig. 5). In all cases, the UPatch successfully acquired the umbilical cord Doppler signals. As a result, the UPatch did not require repositioning and could maintain continuous spectral Doppler sampling at the placental cord insertion throughout the monitoring period.

Supplementary Discussion 21: Ultrasound incident angle

Accurate measurements of blood flow velocities are influenced by the ultrasound incident angle, which is the angle between the ultrasound beam and the direction of blood flow[133]. To derive accurate velocity, measuring the ultrasound incident angle is essential. Conventionally, an angle correction cursor is aligned to the flow direction within the vessel visualized in the duplex image to calculate the incident angle[48,51]. However, this method may be inaccurate because the vessel may not lie in the incident plane[48].

To circumvent the need for angle measurements and eliminate its associated inaccuracies, Doppler indices, such as the systolic-to-diastolic ratio, pulsatility index, and resistance index, are widely employed[48]. These indices are angle-independent because they are normalized (Supplementary Discussion 5)[48]. This allows reliable assessment of vascular resistance and blood flow characteristics that are not subject to the variances of vessel orientations. Consequently, these angle-independent indices are preferred in clinical diagnosis[66,67].

The UPatch operates as a single-beam device using the full aperture to generate either a diverging or focused beam. Future work could enable cross-beam Doppler acquisition, where beams intersect the same vessel segment from different angles to recover the absolute flow velocity[155], and multi-point acquisition, where spatially separated sampling locations allow assessment of pulse wave propagation[156,157]. These capabilities can assess absolute flow or pulse wave velocity values distributed across the feto-placental vessels.

Cross-beam Doppler enables calculation of the blood flow velocity by insonating the same vascular segment with two intersecting ultrasound beams with a defined geometric relationship[155]. The intersecting beams are transmitted using individual transducers[158,159] or electronically steered sub-apertures of an array[160,161], and each beam provides an independent measurement of Doppler frequency shift. Using these values, the blood flow velocity can be calculated as[155]:

$$v = \frac{c}{2F_0}\frac{1}{sin\delta}\sqrt{{f_1}^2 + {f_2}^2 - 2f_1 f_2 cos\delta} \quad (25)$$

where c is the speed of sound, F_0 is the transmit frequency, δ is the intersection angle between the two beams, and f_1 and f_2 are the Doppler frequency shifts.

This calculation yields a flow velocity estimate at the beam intersection with minimal computational power. However, the measurement is highly sensitive to maternal and fetal movements, as even small displacements can cause the beam intersection to no longer coincide with the intended vessel segment. As a result, accurate cross-beam measurements require frequent realignment of the beam intersection with the moving vessel, which has limited the adoption of this technique in dynamic fetal monitoring.

The vessel tracking algorithm developed in this work could provide a pathway toward cross-beam Doppler by continuously updating beam placement to maintain alignment with the target vessel during motion. This approach could enable automated estimation of absolute blood flow velocity and would be particularly advantageous for feto-placental vessels with complex or dynamically changing orientations, such as the umbilical cord.

Multi-point acquisition records pulse waveforms from two or more spatially separated points along a vessel using individual devices, enabling pulse wave velocity measurements[156]. By recording

pulse waveforms at these distinct locations, the system can measure the pulse transit time, the time for the pulse pressure wave to travel between two sites, and calculate the pulse wave velocity[162]:

$$PWV = \frac{L}{PTT} \tag{26}$$

where L is the distance between two recording sites, and PTT is the measured pulse transit time.

As an indicator of vascular tone and stiffness, increased pulse wave velocity in feto-placental vessels reflects compromised placental perfusion and impaired oxygen and nutrient transfer, which are associated with fetal growth restriction and adverse perinatal outcomes. However, accurate measurement of the path length between recording sites is challenging in fetal vessels due to depth, curvature, and continuous motion, leading to uncertainty in pulse wave velocity estimation.

Phase aberration correction in the UPatch beamforming enables accurate image reconstruction across varying maternal abdominal curvatures, preserving geometric fidelity. This allows accurate length estimation of deep feto-placental vessels, after which two beams can be targeted at proximal and distal vessel locations to acquire synchronized pulse waveforms. Combining the measured pulse transit time with the accurately estimated vessel path length enables the precise calculation of pulse wave velocity.

Supplementary Discussion 22: Abnormality in a pre-eclamptic participant

We continuously monitored 52 participants across a range of gestational ages in third trimester and clinical conditions to evaluate the UPatch's ability to capture hemodynamic changes. One pre-eclamptic participant exhibited markedly abnormal patterns in her Doppler indices. To highlight the diagnostic value of continuous monitoring of blood flow, we present a comparative analysis between a healthy participant at 32 weeks 6 days and the pre-eclamptic one at 28 weeks 3 days gestational age.

The average fetal heart rate was 149.5 bpm for the healthy and 143.3 bpm for the pre-eclamptic participants (Fig. 4b). Both were within the normal range of 120–160 bpm (Supplementary Discussion 3)[17,43]. In addition, their fetal heart rate variabilities were 20.3 bpm and 21.5 bpm, respectively, characterized as moderate variability, indicating healthy fetuses (Supplementary Discussion 3)[17,43]. Thus, cardiotocography devices could not detect the compromised fetal health of the pre-eclamptic participant.

Notable differences were observed in the resistance index, pulsatility index, and systolic-to-diastolic ratio between the two participants. The healthy participant had an average resistance index of 0.62, pulsatility index of 1.04, and systolic-to-diastolic ratio of 2.61 (Fig. 4b and Supplementary Fig. 36). In contrast, the pre-eclamptic participant had an average resistance index of 0.85, pulsatility index of 2.04, and systolic-to-diastolic ratio of 5.11 (Fig. 4b and Supplementary Fig. 37).

For the healthy participant's gestational age, the 50th percentile values are 0.63 for resistance index, 0.96 for pulsatility index, and 2.68 for systolic-to-diastolic ratio[68]. For the pre-eclamptic participant's gestational age, the 50th percentile values are 0.66 for resistance index, 1.03 for pulsatility index, and 2.93 for systolic-to-diastolic ratio[68]. The healthy participant's values were

near the 50th percentile, whereas the pre-eclamptic participant's values were markedly higher, indicating poor blood flow exchange between the fetus and placenta.

Following the detection of abnormal umbilical artery blood flow in the pre-eclamptic participant using the UPatch, the research activities were halted. The mother underwent intensive monitoring with clinically validated devices, and the baby was delivered 4 days later, at 203 days of gestation, via Cesarean section. The UPatch detected abnormalities that conventional cardiotocography failed to identify, leading to a timely intervention that saved the fetus. The newborn is thriving, demonstrating the real-world clinical value of this technology.

Supplementary Discussion 23: Clinical importance of continuous monitoring of blood flow

Continuous monitoring of blood flow using the UPatch enables clinically meaningful assessment of fetal well-being throughout pregnancy.

First, in current obstetric practice, cardiotocography is used to continuously monitor fetal heart rate, particularly for inpatients. When prolonged fetal heart rate decelerations or abnormal patterns are observed, Doppler ultrasound is typically employed by sonographers to assess umbilical blood flow[59]. This two-step process is time-consuming and labor-intensive, which may delay or miss the optimal timing to detect fetal complications, reflecting a technological gap. The UPatch addresses this gap by providing continuous monitoring of both fetal heart rate and umbilical blood flow in a single system without the need for a sonographer.

Second, the UPatch enables baseline measurement of umbilical blood flow and differentiate normal from abnormal physiological fluctuations. Normal fluctuations can temporarily alter spectral Doppler waveforms. When captured by intermittent scans, such altered values do not reflect true pathology and may raise false positive concerns or trigger unnecessary clinical interventions (Supplementary Figs 36–87)[163]. This limitation of intermittent scans reduces reliability in dynamic and complicated clinical settings, often requiring serial measurements over several days. The UPatch addresses this limitation by providing a continuous data stream of umbilical blood flow, allowing clinicians to distinguish sustained compromise from transient, self-resolving physiological fluctuations, which supports more reliable, real-time clinical decision-making.

Third, unlike fetal heart rate, which requires complex time-series analysis, Doppler indices offer clinically validated measurements of fetal hemodynamics[9,164]. Fetal heart rate interpretation relies on pattern recognition to assess baseline, variability, accelerations, and decelerations over a long period of time, because isolated values are insufficient to reveal fetal well-being[9]. However, these time-series features show poor correlation with gestational trends and perinatal risks[2,9,165]. In contrast, the three Doppler indices used in this study are well-established parameters heavily used in hospitals to assess placental and fetal vascular resistance[66-68,166,167]. Doppler indices have well-defined reference ranges, exhibit consistent gestational trends, and stratify risk groups with great sensitivity[59,67,68]. Continuous monitoring with the UPatch enables real-time acquisition of these indices without the need for time-series analysis (Fig. 4e–i and Supplementary Figs 93–95).

Fourth, the UPatch provides continuous, actionable information that supports more precise risk management of the trade-off between preterm delivery (delivery before 37 weeks) and stillbirth than cardiotocography[168]. Before 37 weeks, clinicians aim to avoid delivery unless fetal

compromise is evident. In a pre-eclamptic patient (28 weeks 3 days gestational age) in this study, the UPatch identified umbilical flow abnormalities that were not evident on conventional cardiotocography, prompting an intervention that saved a baby (Fig. 4a–d).

Supplementary Discussion 24: Data processing

Data preprocessing and processing were performed locally on the backend system (Vantage 256, Verasonics), which is a research-grade ultrasound platform widely used for real-time data acquisition[169,170]. Data preprocessing is the initial step, including down-sampling and denoising, which are applied to the raw radiofrequency signals. Then, data processing, including beamforming to generate duplex images, autonomous tracking to identify the sample gate, and spectral Doppler estimation to extract velocity waveforms, is implemented for real-time data acquisition.

By directly transferring raw radiofrequency signals to the backend system, this protocol ensured minimal latency with negligible delay. Privacy was maintained because all participant data were transmitted by a flexible cable and remained within the local processing station, eliminating security concerns associated with wireless transmission or cloud-based storage. Additionally, the study was approved by the Institutional Review Board, ensuring compliance with privacy regulations. Computational requirements were sufficiently managed by the backend system, which was specifically designed to handle high-throughput ultrasound data processing[170].

To optimize storage, the system prioritizes clinically actionable data and discard unnecessary high-volume raw imaging. Specifically, instead of storing full duplex images and spectral Doppler waveforms, the system extracts and saves Doppler indices (such as resistance index, pulsatility index, and systolic-to-diastolic ratio) and fetal heart rate, which are one-dimensional and require minimal storage. This approach ensures storage efficiency while providing essential information for clinical decision-making.

Supplementary Discussion 25: Predictive modeling

To assess the feasibility of risk classification using UPatch measurement features, we performed pilot predictive modeling using participant-level mean Doppler indices and fetal heart rate (Supplementary Fig. 96)[171]. While limited sample sizes within individual perinatal condition groups may constrain predictive performance, we piloted implementing logistic regression to evaluate whether mean fetal heart rate, pulsatility index, systolic-to-diastolic ratio, and resistance index could distinguish perinatal risk conditions[172].

Receiver operating characteristic (ROC) analyses were used to evaluate the ability of a predictive measurement to quantify discriminative performance among different health outcomes[173]. The x-axis shows the false positive rate (1 – specificity; specificity quantifies the proportion of true negatives correctly identified), and the y-axis shows the true positive rate (sensitivity; sensitivity measures the proportion of true positives correctly identified). Each point on the curve represents the performance of the classifier at a different decision threshold, which is the cutoff to convert a predicted probability into a perinatal risk condition. The area under the ROC curve (AUROC) provides a single summary metric of discriminative performance: a value of 0.5 corresponds to

chance-level performance (no predictive ability), whereas values closer to 1.0 indicate stronger ability to distinguish perinatal risk conditions. By comparing AUROC values for fetal heart rate and Doppler indices, we can quantify how well each measurement predicts perinatal risk.

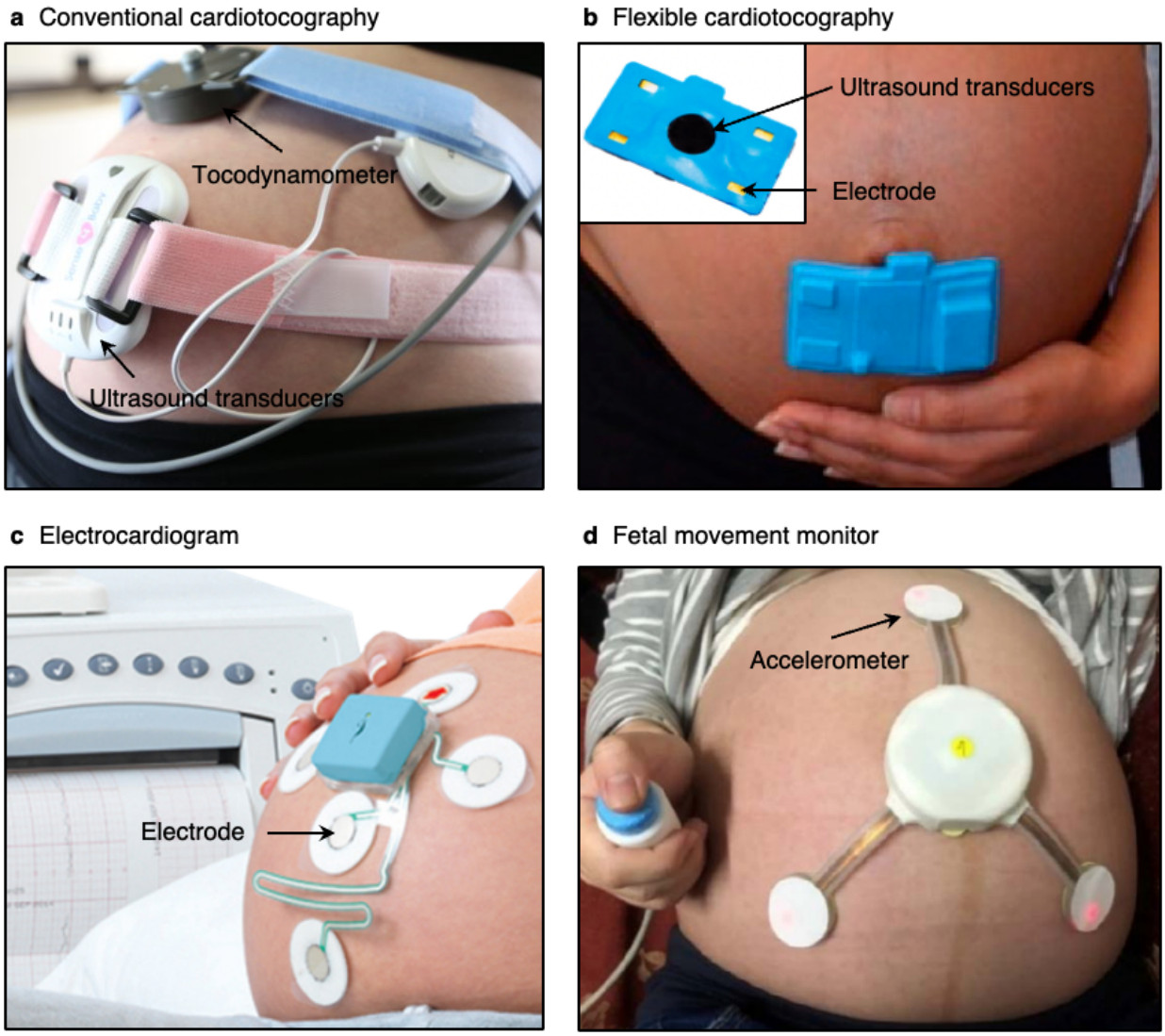
For the healthy, large for gestational age, pre-eclampsia, and hypertension groups, AUROC values for Doppler indices were in the range of 0.605–0.651 compared with 0.533–0.646 for fetal heart rate, reflecting modest but systematic improvement in discrimination. For the small for gestational age group, Doppler indices achieved AUROC values in the range of 0.959–0.962, substantially exceeding the corresponding fetal heart rate AUROC of 0.511, indicating markedly stronger discriminative performance. For the diabetes group, classification performance remained near chance level for both Doppler indices and FHR with AUROC values of 0.522–0.525 and 0.562, respectively, suggesting limited separability.

Supplementary Discussion 26: Circuit integration

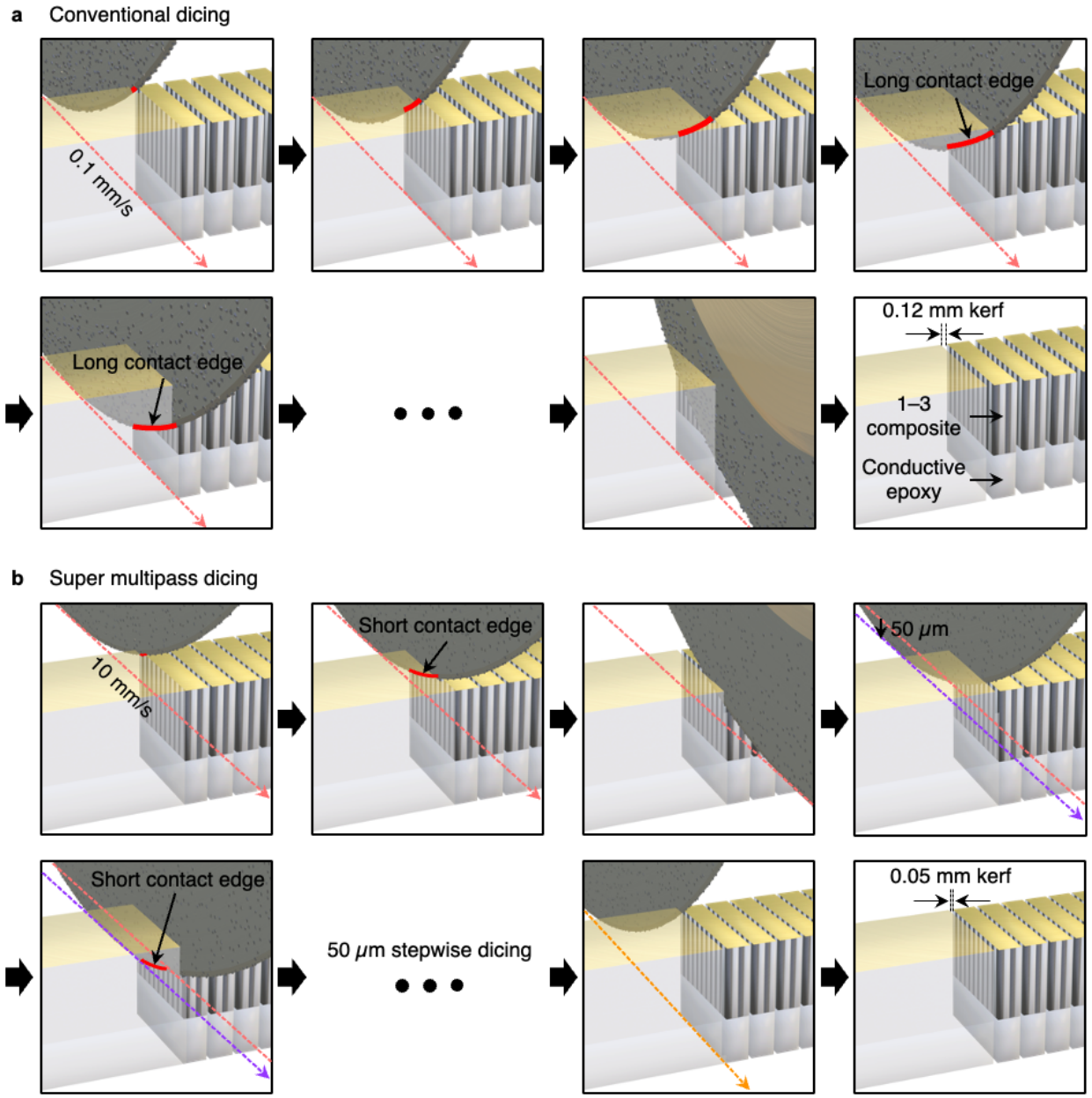
The current UPatch with a wired connection for power and data is optimal for hospital monitoring. However, wireless home monitoring presents distinct clinical advantages. First, it offers a comfortable environment that can yield more precise baseline measurements of fetal blood flow, which is critical for high-risk pregnancies (such as gestational diabetes and pre-eclampsia)[54]. Additionally, it can allow research into how maternal exercise influences fetal hemodynamics[55]. Second, continuous home monitoring generates more extensive data over time, allowing for trend analysis and research into the mechanisms of fetal complications and stillbirth[174]. Third, it supports regular assessments without the inconvenience and costs associated with hospital visits, crucial for those with limited access to healthcare facilities, especially in low/middle-income areas[11,175].

A recently introduced wearable ultrasound circuit has enabled wireless home monitoring[134]. This system can multiplex the signals from all 32 channels of transducer elements through a single transceiver. A Wi-Fi module transmits the received signals to a smartphone, monitoring tissue interfaces and providing critical insights during physical activities (such as biking and high-intensity training). However, its imaging capabilities are limited due to the circuit comprising only a single transceiver. First, while multiplexing can enable imaging, the transducer elements are activated sequentially. This means that there is a time delay between received signals, resulting in low image quality from moving tissue interfaces and red blood cells[176]. Second, for color Doppler imaging, the transceiver must activate the same transducer at a high pulse repetition frequency (~3000 Hz) to assess the mean phase differences among multiple pulse echos[85]. This setup, while feasible for a single transducer, results in a loss of spatial resolution and intrinsically low Doppler signal-to-noise ratio[86]. Due to these limitations, this circuit is inadequate for producing high-quality ultrasound images.

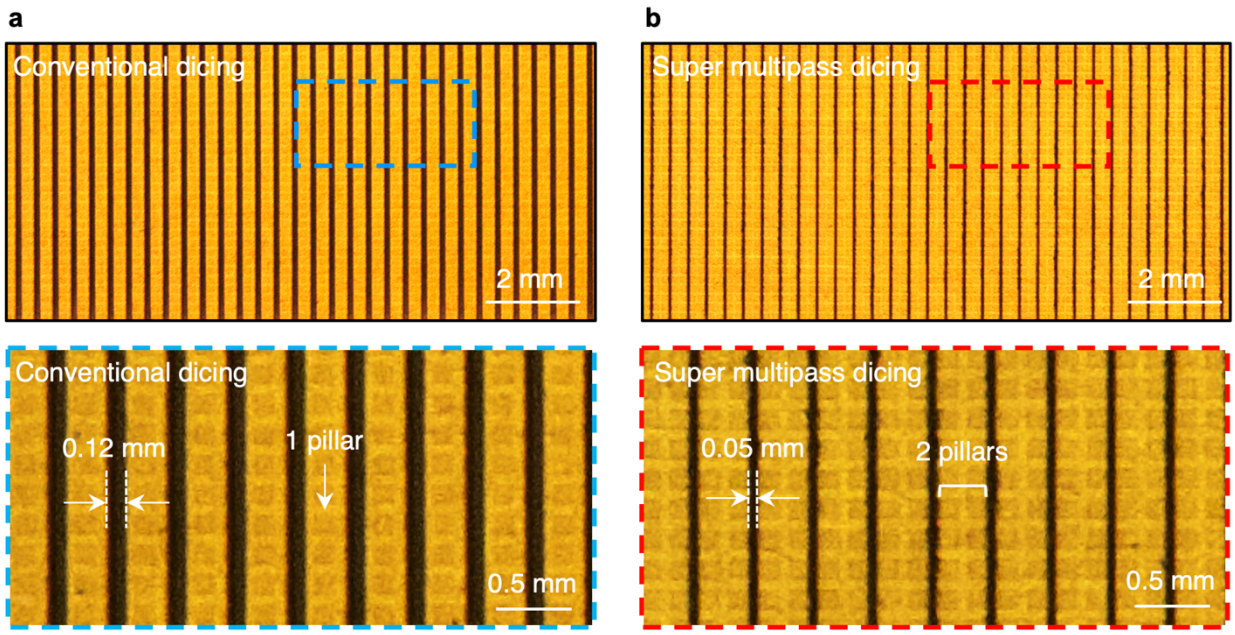
Multiple transceivers are needed to achieve high-quality imaging, but there are three major challenges[177]. First, the increased power consumption from additional transceivers will necessitate more efficient power management solutions to sustain battery life[178]. Second, heat generation poses a substantial challenge because additional circuit components will cause heat accumulation, requiring advanced thermal management strategies to ensure the device's safe use[177]. Third, multiple transceivers will complicate the firmware design, requiring more sophisticated signal processing algorithms[177]. Despite its challenges, the integration of multiple transceivers into wearable ultrasound circuits will enable high-quality imaging for wireless home monitoring.



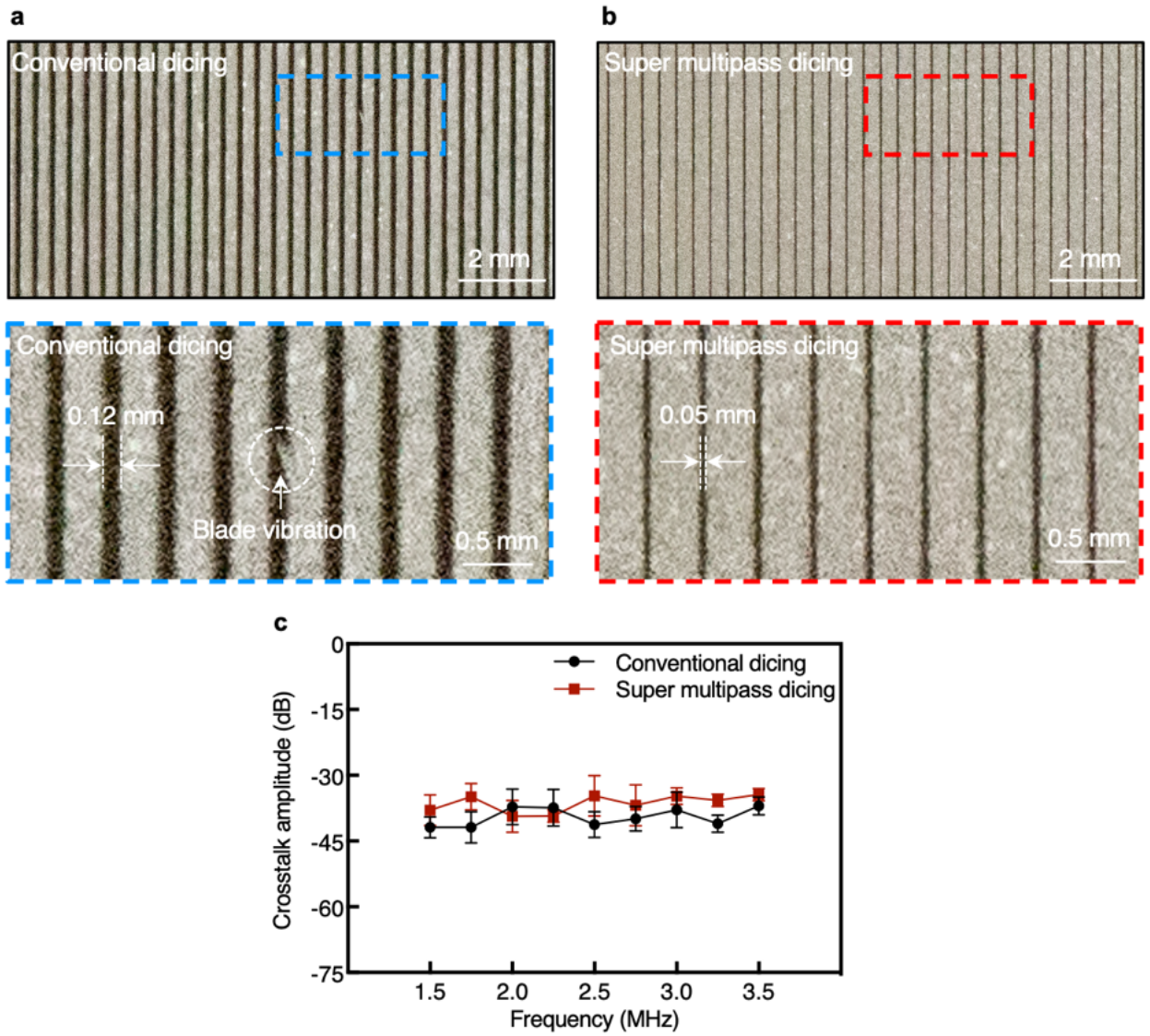
Supplementary Fig. 1 | Existing wearable fetal monitors. a, Conventional cardiotocography comprising ultrasound transducers for fetal heart rate monitoring and a tocodynamometer for uterine contraction monitoring developed by Sense4baby[20,179]. **b**, Flexible cardiotocography using two ultrasound transducers for fetal heart rate recording and four electrodes for maternal electrocardiogram and electrohysterogram to derive uterine contraction developed by Sibel Health[23]. **c**, Fetal electrocardiogram measured with five electrodes to derive fetal heart rate developed by Monica Healthcare (now GE Healthcare)[25,180]. This configuration claims to derive fetal heart rate regardless of fetal position in the maternal abdomen. **d**, Fetal movement sensor using accelerometers[38].



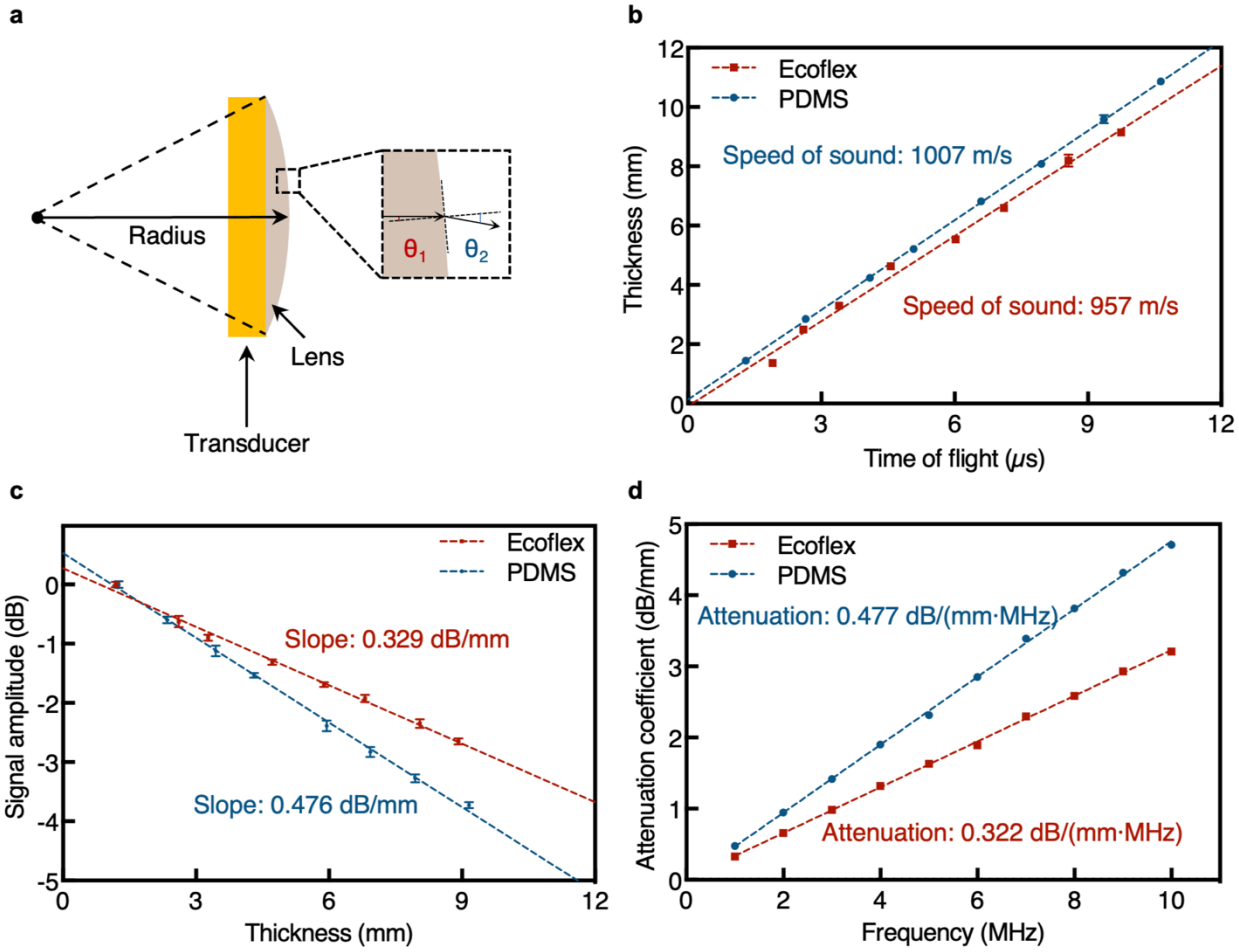
Supplementary Fig. 2 | Dicing strategies. a, Conventional dicing strategy. The blade is lowered to the entire transducer depth of 1 mm and diced at 0.1 mm/s feed speed. This method results in a long contact edge, producing a 0.12 mm kerf. The entire process takes ~15 h to dice a 64-channel array with a 0.4 mm pitch for a 25.55 mm by 12 mm transducer element. **b**, Super multipass dicing used in this study. A thinner dicing blade is used with 50 µm stepwise depth cuts. The blade is first lowered 50 µm from the top of the transducer and diced at 10 mm/s. Then, it is subsequently lowered to another 50 µm depth for dicing. This method results in a short contact edge. A total of 20 stepwise cuts are performed to achieve a 0.05 mm kerf for the same array size, which only takes ~3 h.



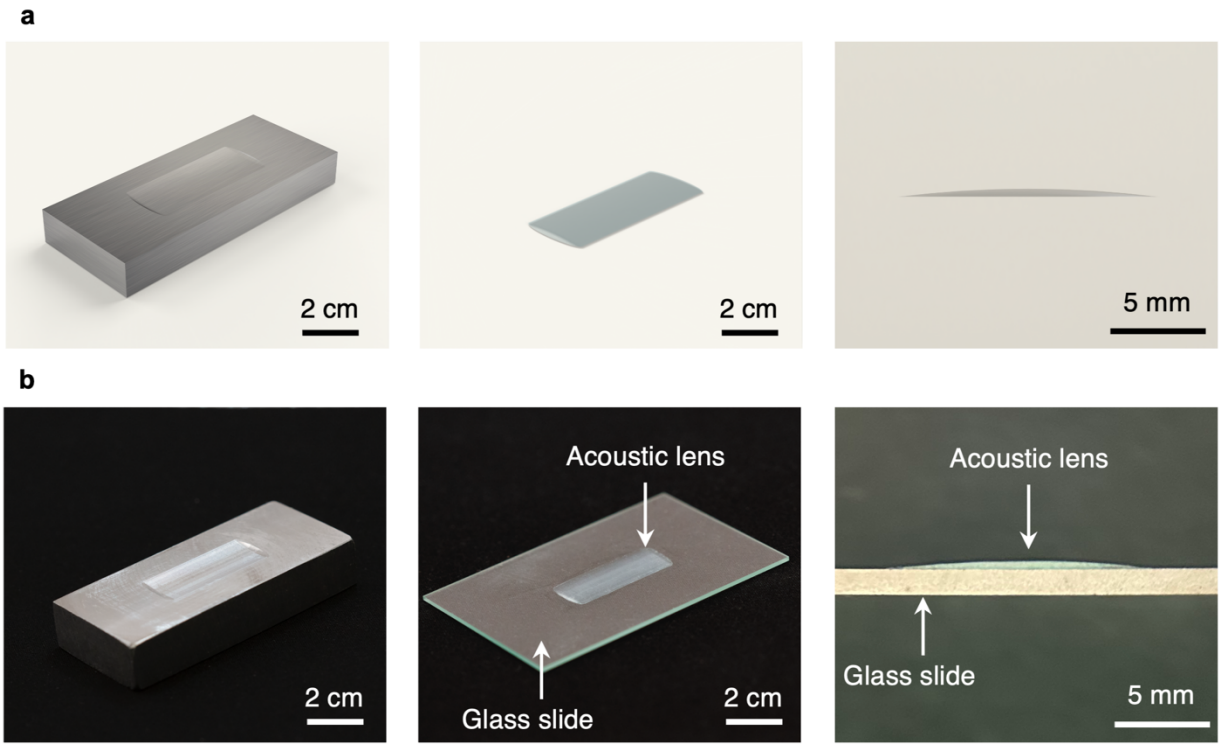
Supplementary Fig. 3 | 1–3 composite side of the transducers diced with different strategies. **a**, Optical images of transducers diced with the conventional strategy. The kerf is 0.12 mm and can only accommodate one piezoelectric pillar of the 1–3 composite. **b**, Optical images of transducers diced with the super multipass strategy. The kerf is 0.05 mm and can accommodate two piezoelectric pillars of the 1–3 composite.



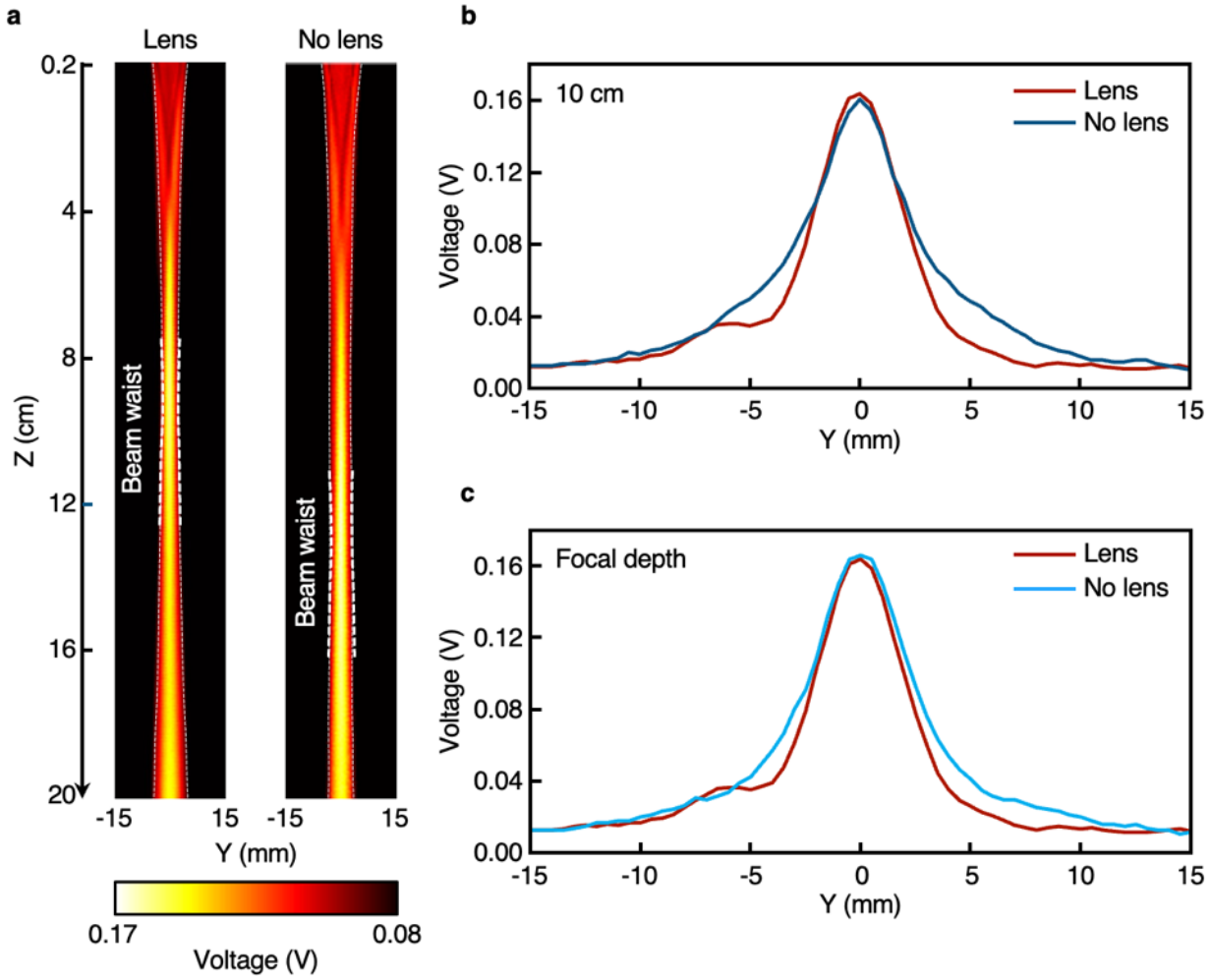
Supplementary Fig. 4 | Comparison of transducers diced with different strategies. a, Optical images of the backing layer side of the transducer diced with the conventional strategy. The kerf is 0.12 mm and blade vibration can cause defects in the transducers. **b**, Optical images of backing layer side of the transducer diced with the super multipass strategy. The kerf is 0.05 mm. **c**, Transducers diced with both strategies demonstrate similar crosstalk amplitude. Data in **c** are presented as mean ± standard deviations (n = 4).



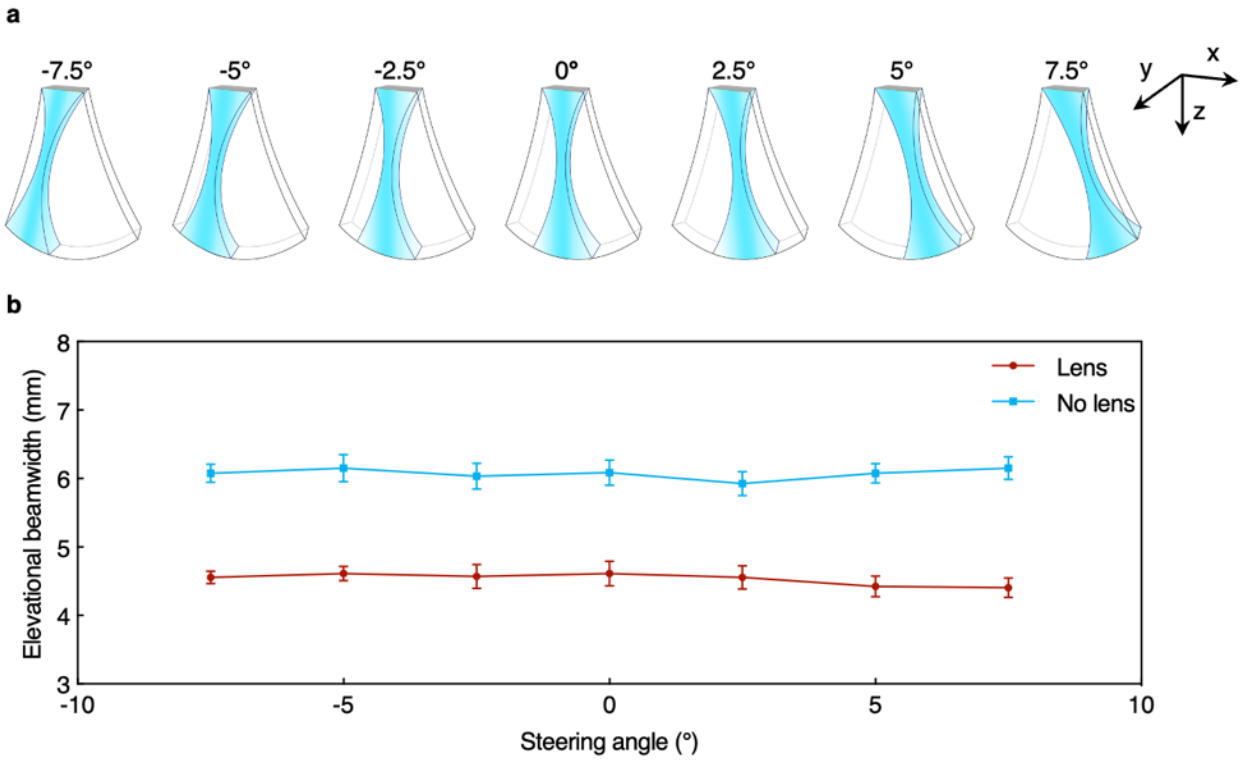
Supplementary Fig. 5 | Design of the acoustic lens. a, Schematics of the lens. The speed of sound of the lens material is used to determine the lens radius. θ_1 and θ_2 are the incidence angle and refraction angle, respectively. **b**, Speed of sound and **c**, acoustic attenuation coefficients at 1 MHz ultrasound frequency of soft acoustic lens materials. **d**, Acoustic attenuation coefficients at 1–10 MHz ultrasound frequencies of soft acoustic lens materials. Each data point in **d** is obtained from a linear fit in **c**. Ecoflex has a lower speed of sound and attenuation coefficient than those of polydimethylsiloxane (PDMS). Data in **b** and **c** are presented as mean ± standard deviations (n = 4).



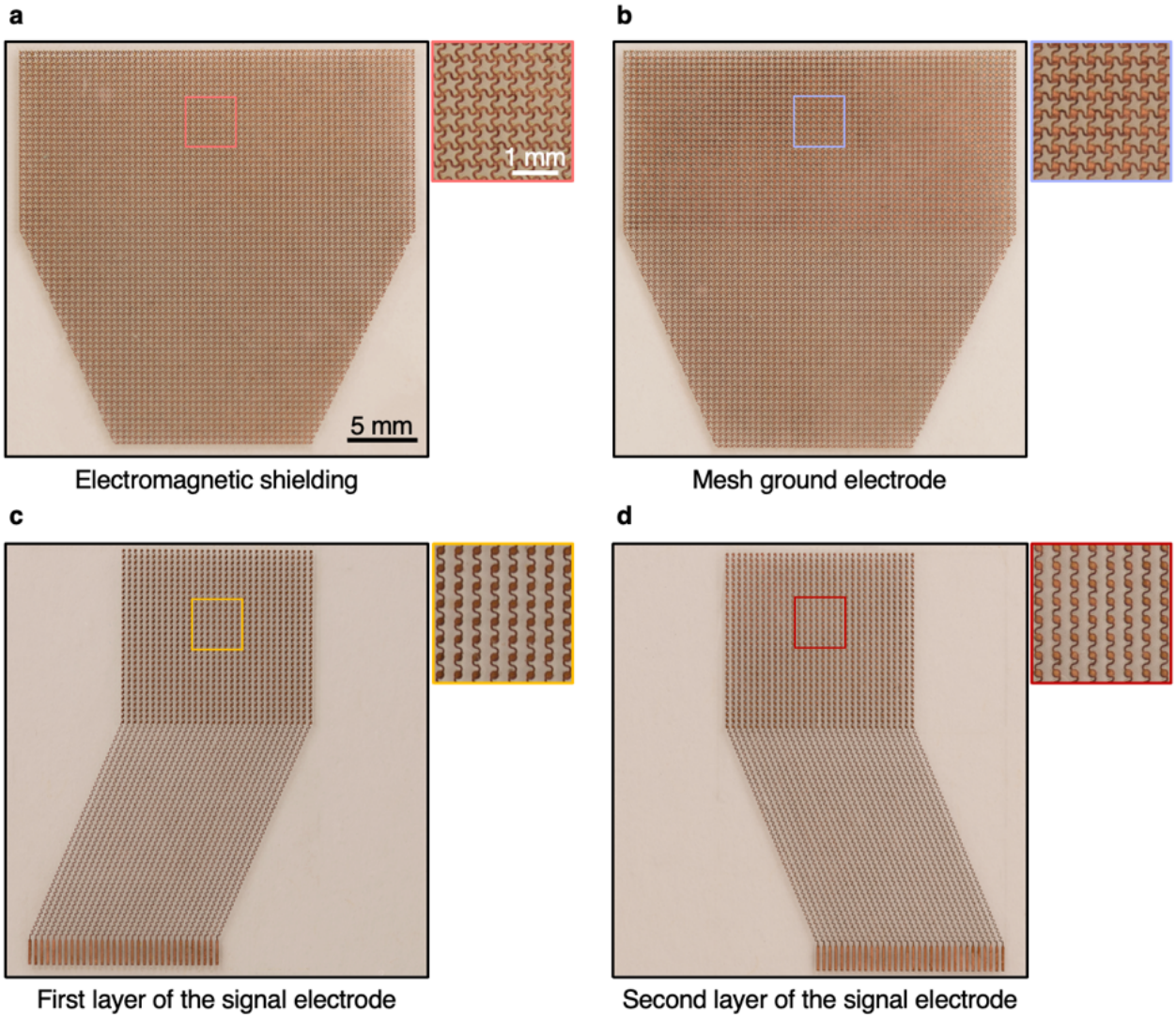
Supplementary Fig. 6 | Fabrication of the acoustic lens. a, Designs and **b**, optical images of the aluminum block mold (left), the molded lens (middle), and the cross section of the lens (right). The aluminum block mold was fabricated using computer numerical control machining. Ecoflex was molded and cured inside the mold to produce an acoustic lens.



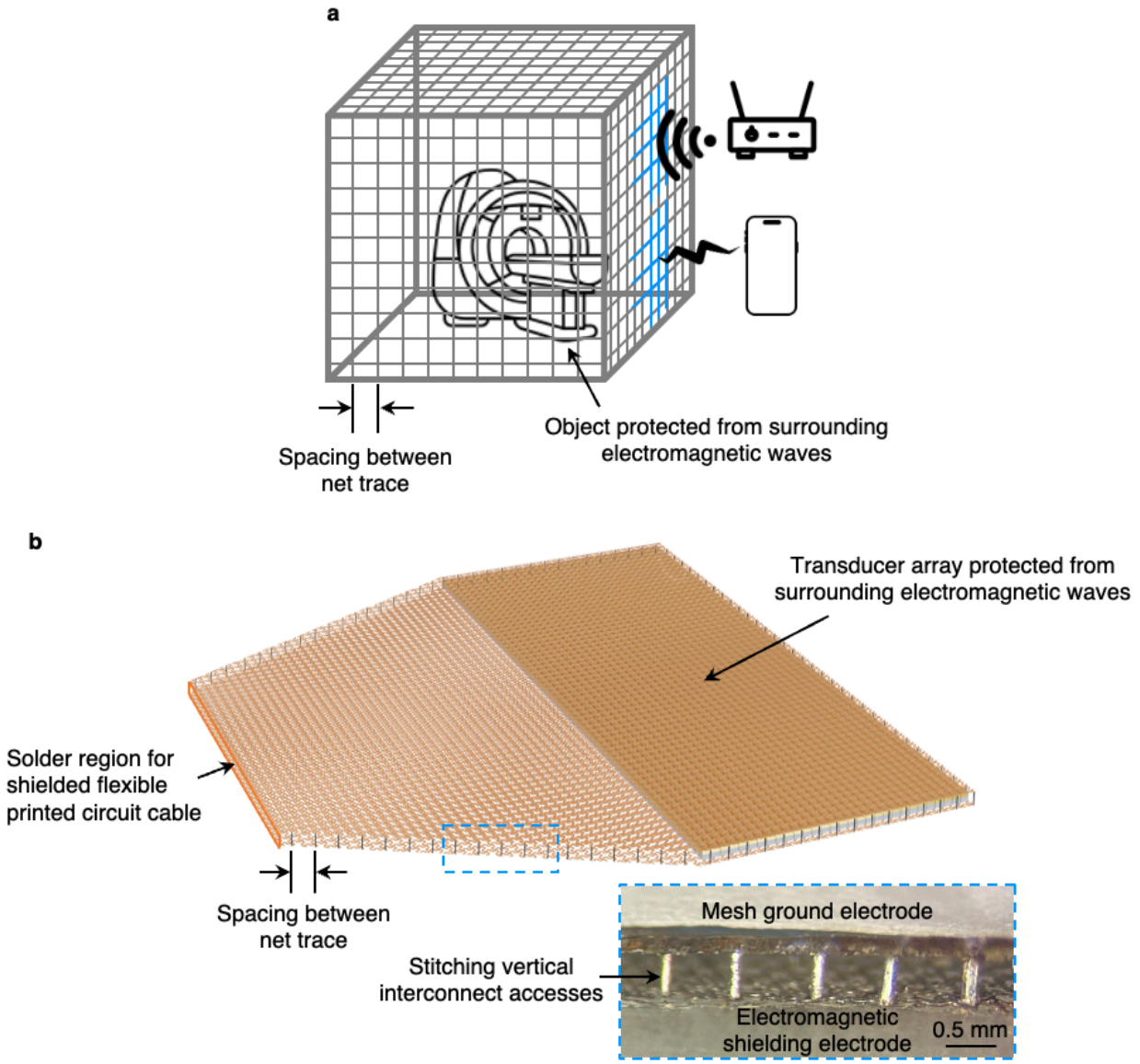
Supplementary Fig. 7 | Characterization of the acoustic lens. a, Hydrophone scanning of the acoustic fields of the UPatch with and without the lens in the YZ plane. The acoustic lens shifts the elevational focal region from ~14 cm to ~10 cm, the typical depth of interest for fetal monitoring. The images share the same color bar for hydrophone output voltage, which can be converted to acoustic pressure. Elevational beamwidth comparisons at **b**, 10 cm and **c**, focal depth with and without the acoustic lens. For **c**, the red line is the signal at 10 cm (lens) while the light blue line is the signal at 14 cm (no lens). In both **b** and **c**, the lens decreases the elevational beamwidth with negligible attenuation.



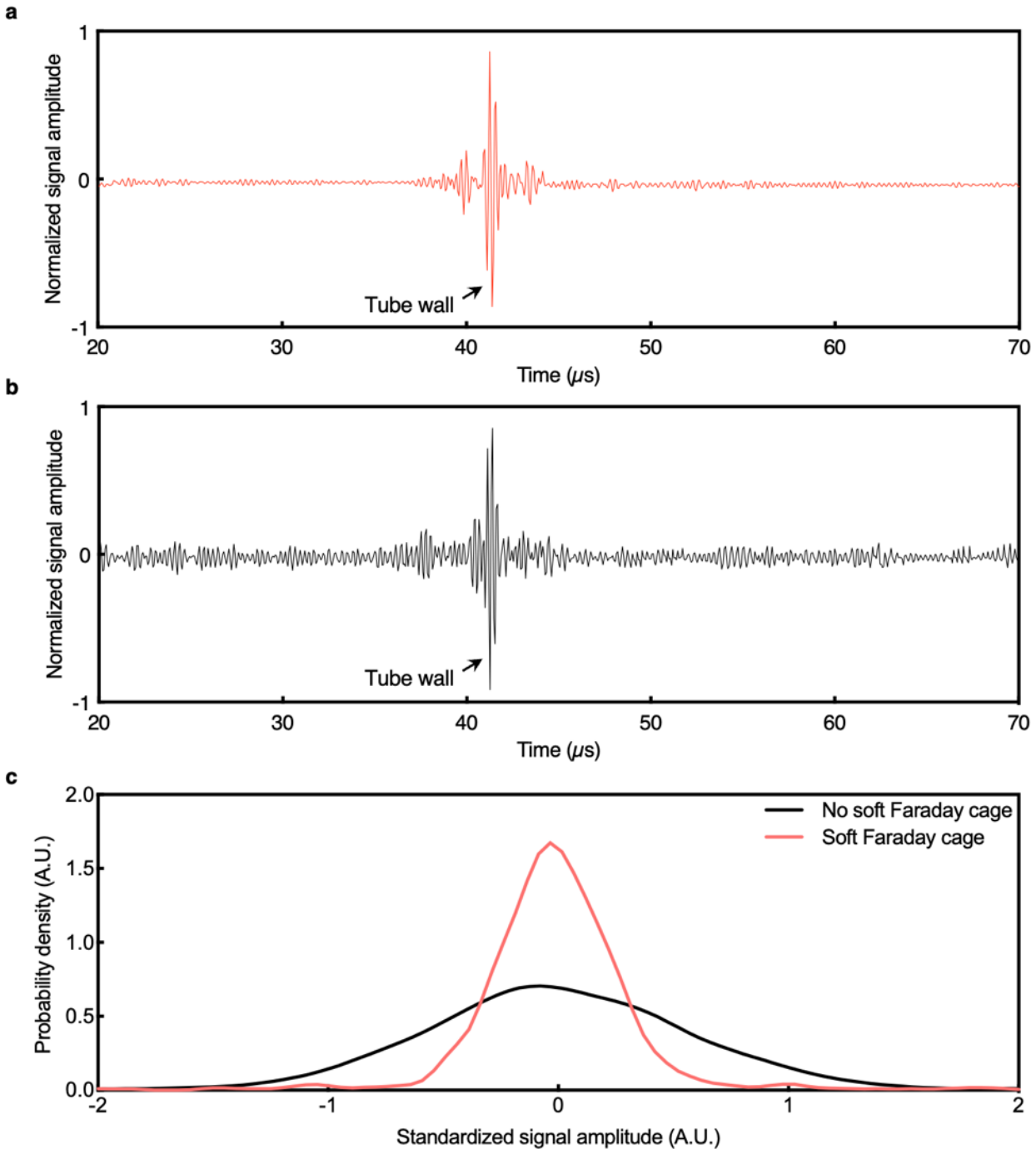
Supplementary Fig. 8 | Elevational beamwidth during beam steering. a, Schematics illustrating three-dimensional acoustic propagation during beam steering. Beam steering occurs in the XZ plane, while the acoustic lens focuses the beam in the YZ plane. **b**, Experimental data comparing the elevational beamwidth at different steering angles with and without the acoustic lens. In both cases, the elevational beamwidth remains consistent across the steering range. Data in **b** are presented as mean ± standard deviations (n = 4).



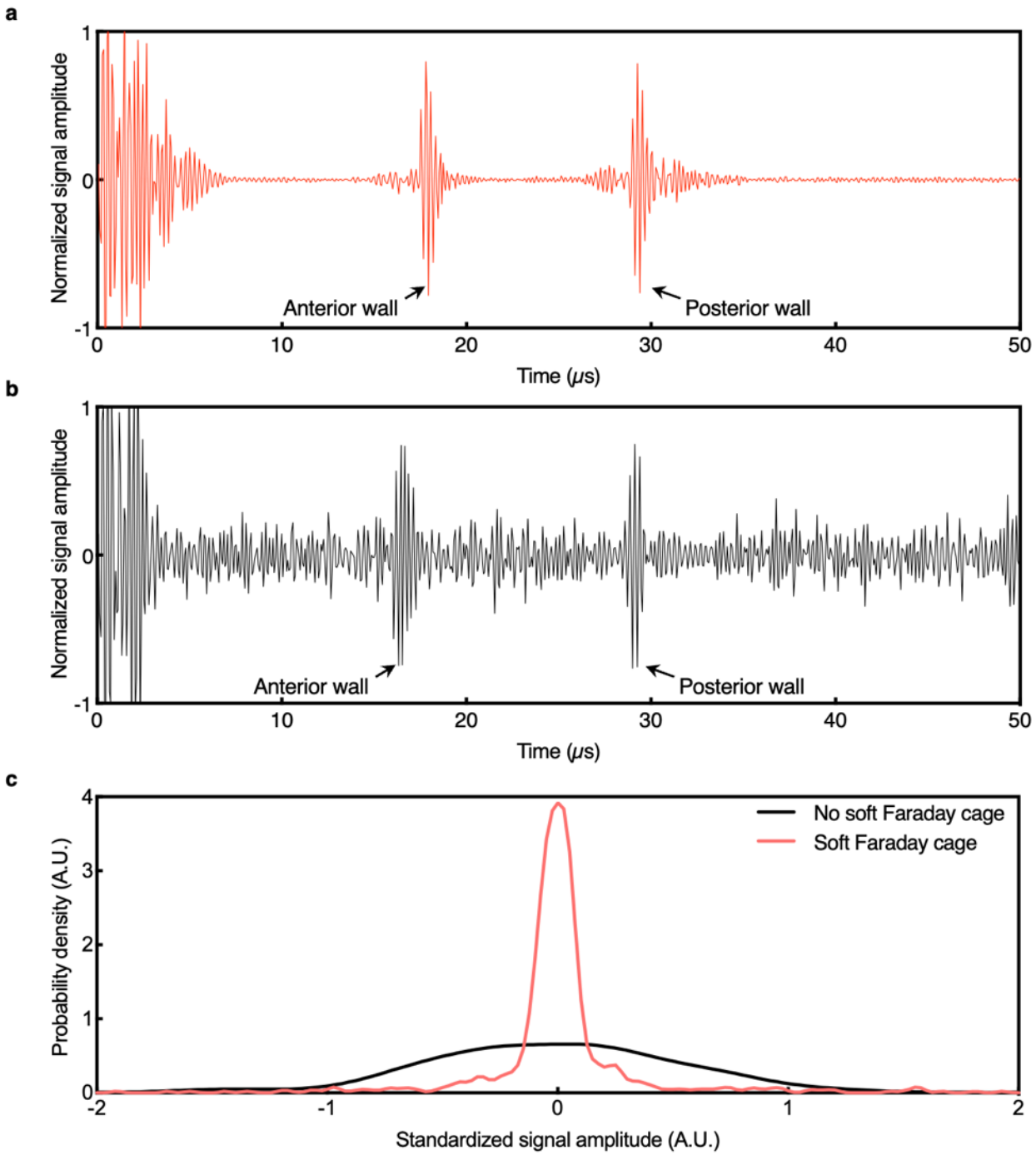
Supplementary Fig. 9 | Photographs of each layer of electrodes. a, Electromagnetic shielding layer. **b**, Mesh ground electrode layer. **c**, The first layer of the signal electrode. **d**, The second layer of the signal electrode. Insets are the serpentine design of each layer. The first and second layers of the signal electrodes activate 32 channels each. The electromagnetic shielding and mesh ground electrode layers minimize electromagnetic interference from the surrounding environment. The photographs and insets share the same scale bar, respectively.



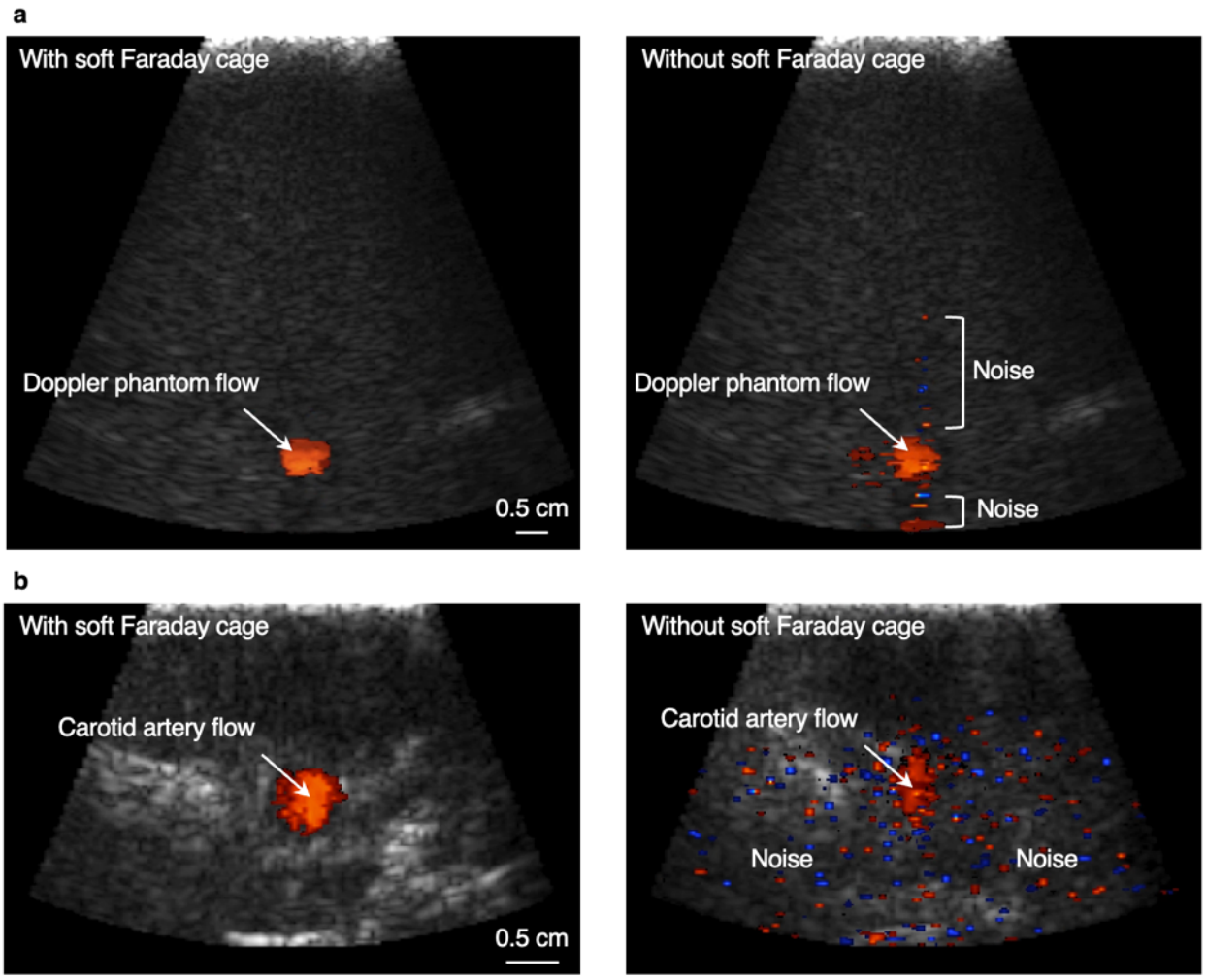
Supplementary Fig. 10 | Faraday cages. a, Schematics of a Faraday cage. The Faraday cage protects sensitive electronic equipment by absorbing electromagnetic waves from other electronic sources (e.g., Wi-Fi routers and smartphones). The spacing between the metal traces determines the frequency bands of the electromagnetic waves that can be blocked. **b**, Schematics of the soft Faraday cage. Inset is a zoomed-in photograph of the stitching vertical interconnect accesses. The electromagnetic shielding and mesh ground electrode layers are patterned with serpentine traces and encapsulated in a soft elastomer. The stitching vertical interconnect accesses enable a three-dimensional enclosure of the transducer, thereby providing effective isolation within the soft Faraday cage.



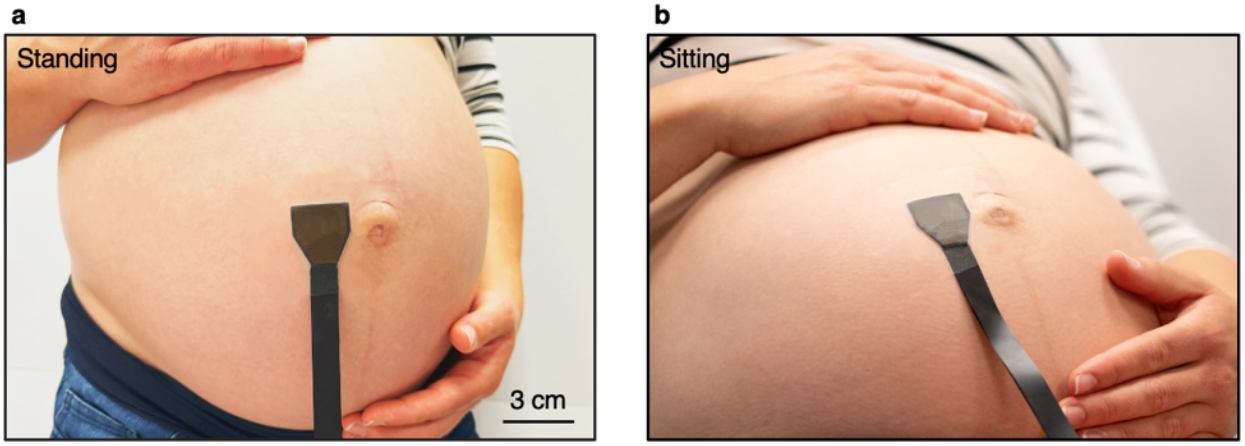
Supplementary Fig. 11 | Effects of the soft Faraday cage on phantom measurements. Radiofrequency signals **a**, with and **b**, without the soft Faraday cage. **c**, Probability density function comparison with and without the soft Faraday cage. The narrower distribution of measurements with the soft Faraday cage indicates reduced susceptibility to ambient electromagnetic noise.



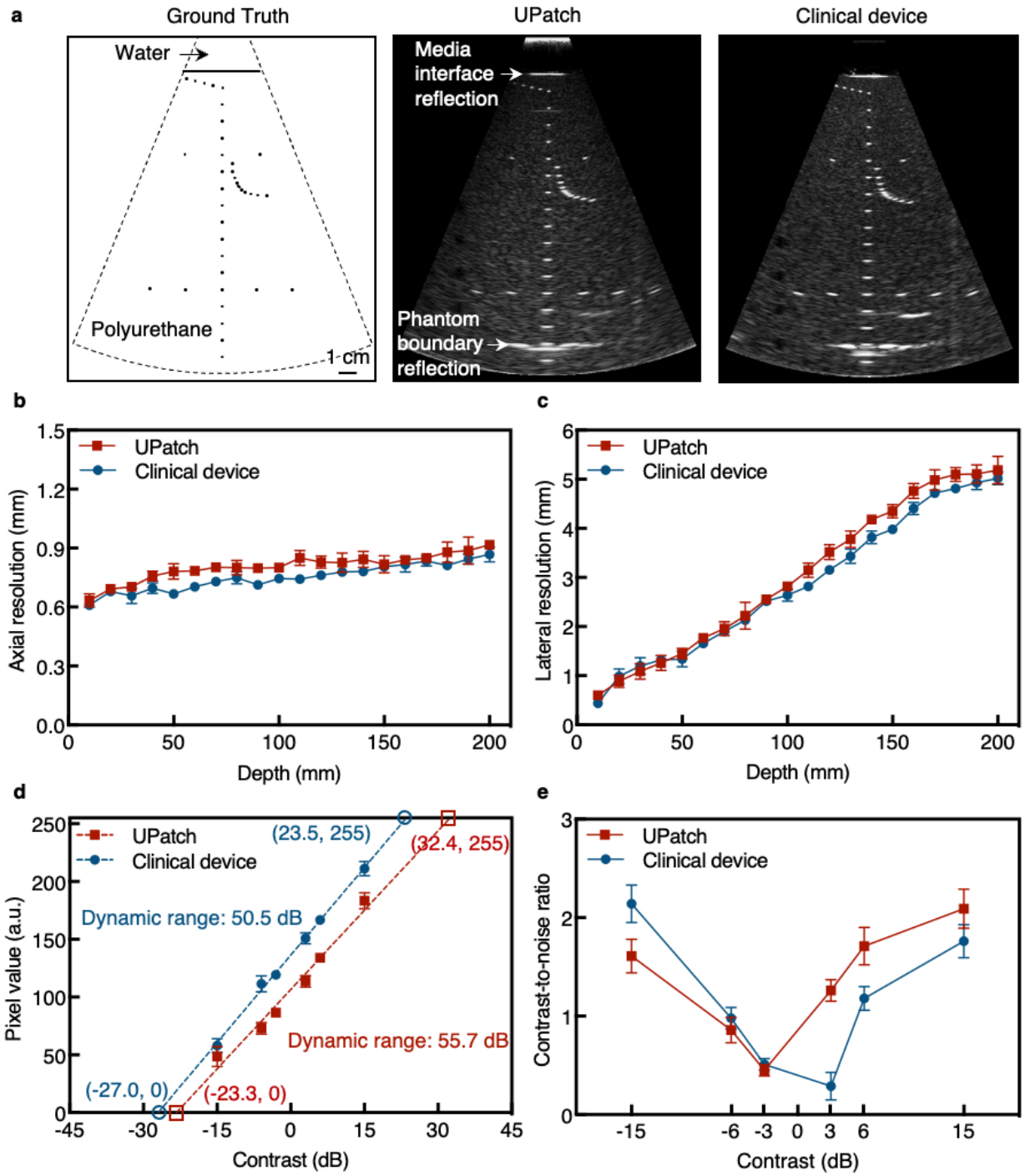
Supplementary Fig. 12 | Effects of the soft Faraday cage on adult carotid artery measurements. Radiofrequency signals **a**, with and **b**, without the soft Faraday cage. **c**, Probability density function comparison with and without the soft Faraday cage. The narrower distribution of measurements with the soft Faraday cage indicates reduced electromagnetic interference, potentially from both physiological and environmental sources, enhancing in vivo signal fidelity.



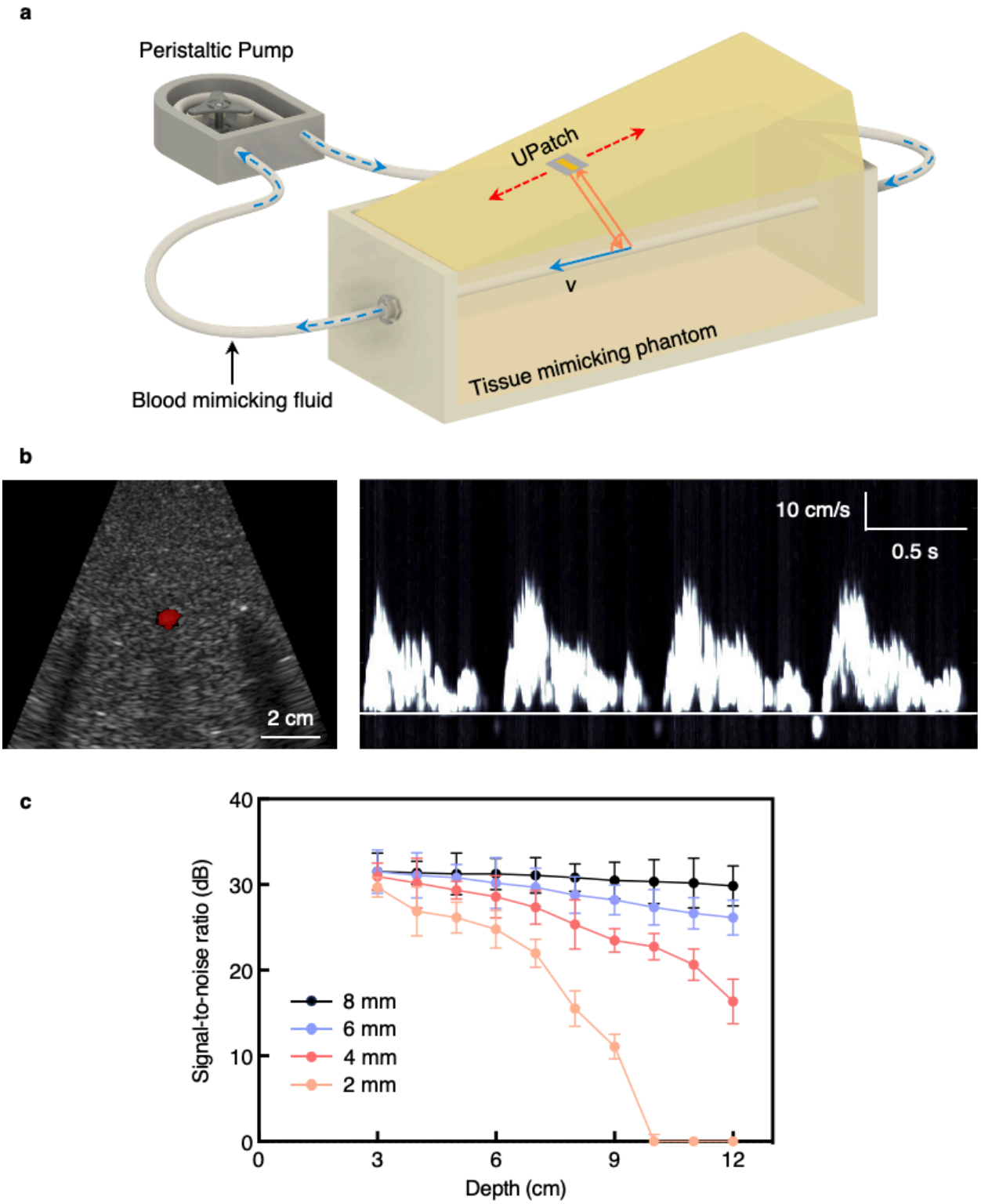
Supplementary Fig. 13 | Effects of the soft Faraday cage on noise levels in duplex imaging. a, Duplex images of a phantom with and without the soft Faraday cage. The images share the same scale bar. **b**, Duplex images of the carotid artery in a healthy adult with and without the soft Faraday cage. The images share the same scale bar. The use of the soft Faraday cage reduces noise and improves image quality.



Supplementary Fig. 14 | Photographs of the UPatch on a pregnant woman. The UPatch is laminated on a pregnant participant in **a**, standing and **b**, sitting positions. The UPatch conforms closely to the contour of the abdominal surface, ensuring intimate contact for effective ultrasound transmission. The photographs share the same scale bar.

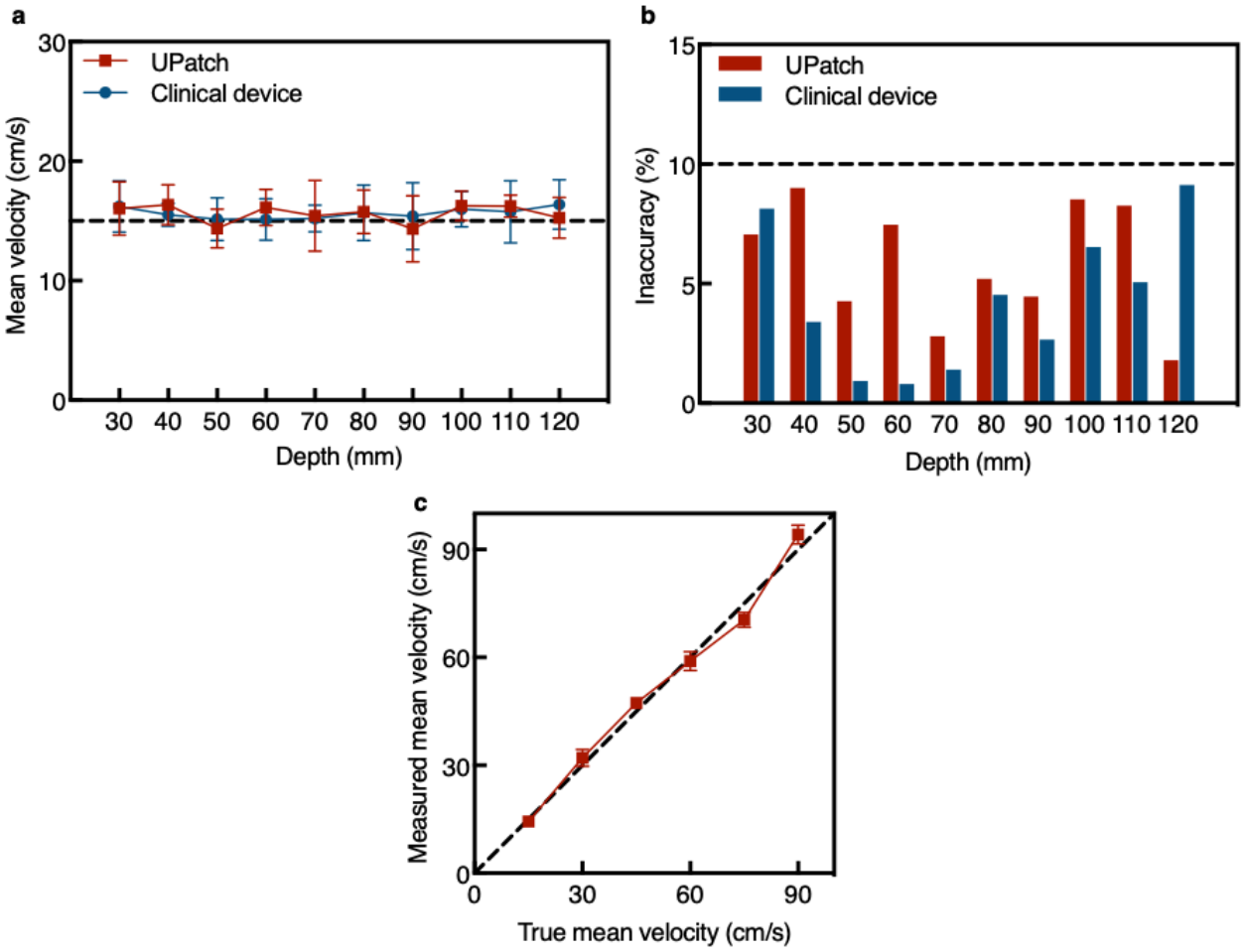


Supplementary Fig. 15 | Brightness mode characterizations. **a**, Schematics of the phantom (left), and brightness-mode images from the UPatch (middle) and a clinical device (P4–1, ATL) (right). The brightness-mode image quality is comparable between the UPatch and clinical device. The images share the same scale bar. **b**, Axial resolution from depths 10 mm to 200 mm. **c**, Lateral resolution from depths 10 mm to 200 mm. **d**, Dynamic range calculated based on six grayscale targets (-15 dB, -6 dB, -3 dB, +3 dB, +6 dB, and +15 dB). **e**, Contrast-to-noise ratio calculated based on the six grayscale targets. Data in **b-e** are presented as mean ± standard deviations (n = 4).

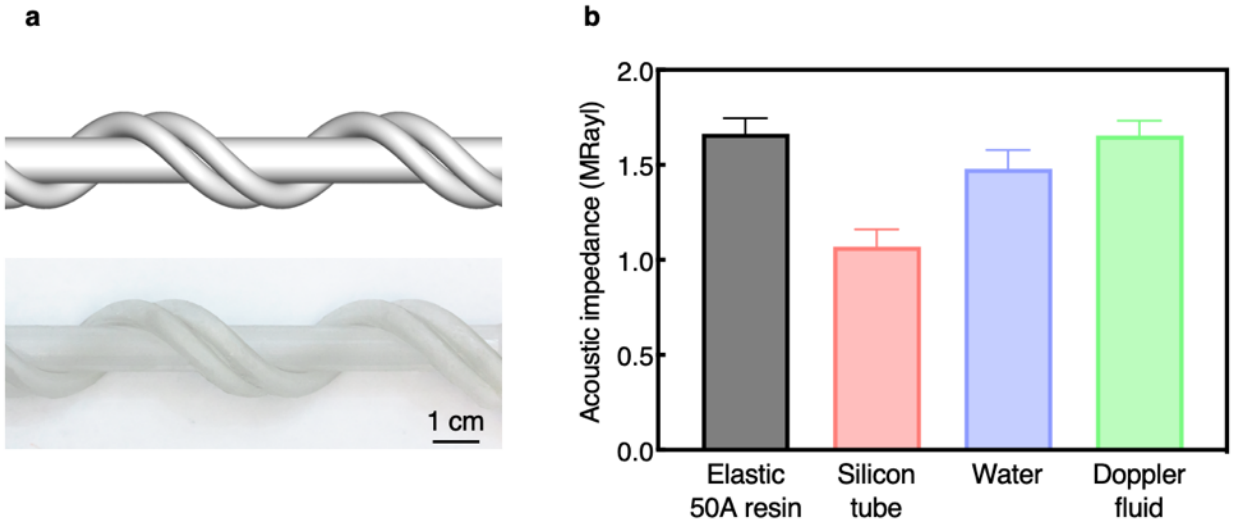


Supplementary Fig. 16 | Doppler characterizations. **a**, Schematic setup of a commercial Doppler phantom (CIRS ATS523A) for Doppler characterizations. Blood-mimicking fluid was pumped into the phantom with a known Doppler angle using a peristaltic pump. The depth from the UPatch to the phantom vessel ranges from 3 cm to 12 cm. **b**, Exemplary color-flow image (left) and spectral Doppler signal (right) acquired from the phantom using the UPatch. **c**, Signal-to-noise

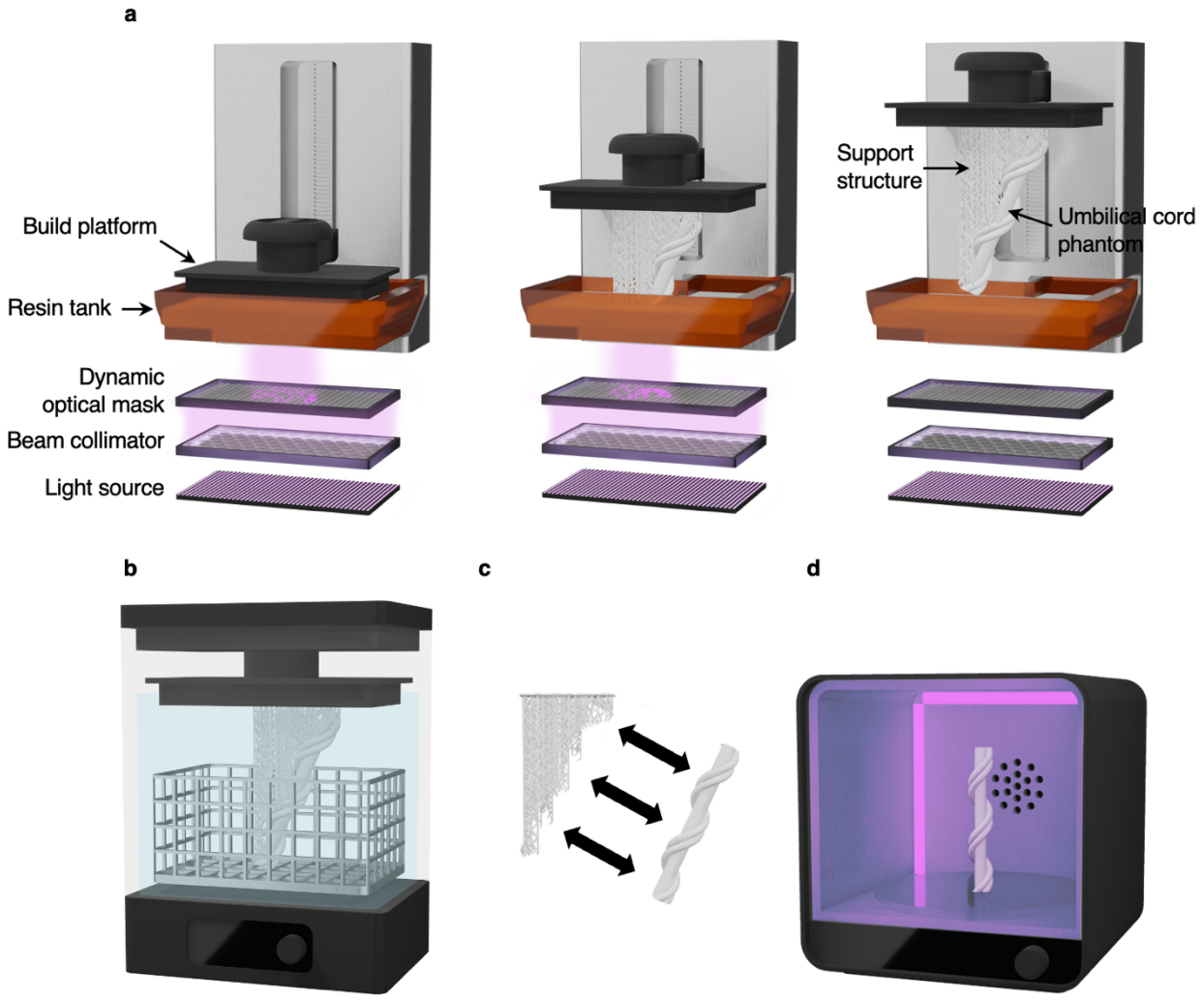
ratio of color Doppler signals of the UPatch at different phantom vessel depths. The UPatch can detect the smallest phantom vessel diameter of 2 mm at a depth of 10 cm. Data in **c** are presented as mean ± standard deviations (n = 4).



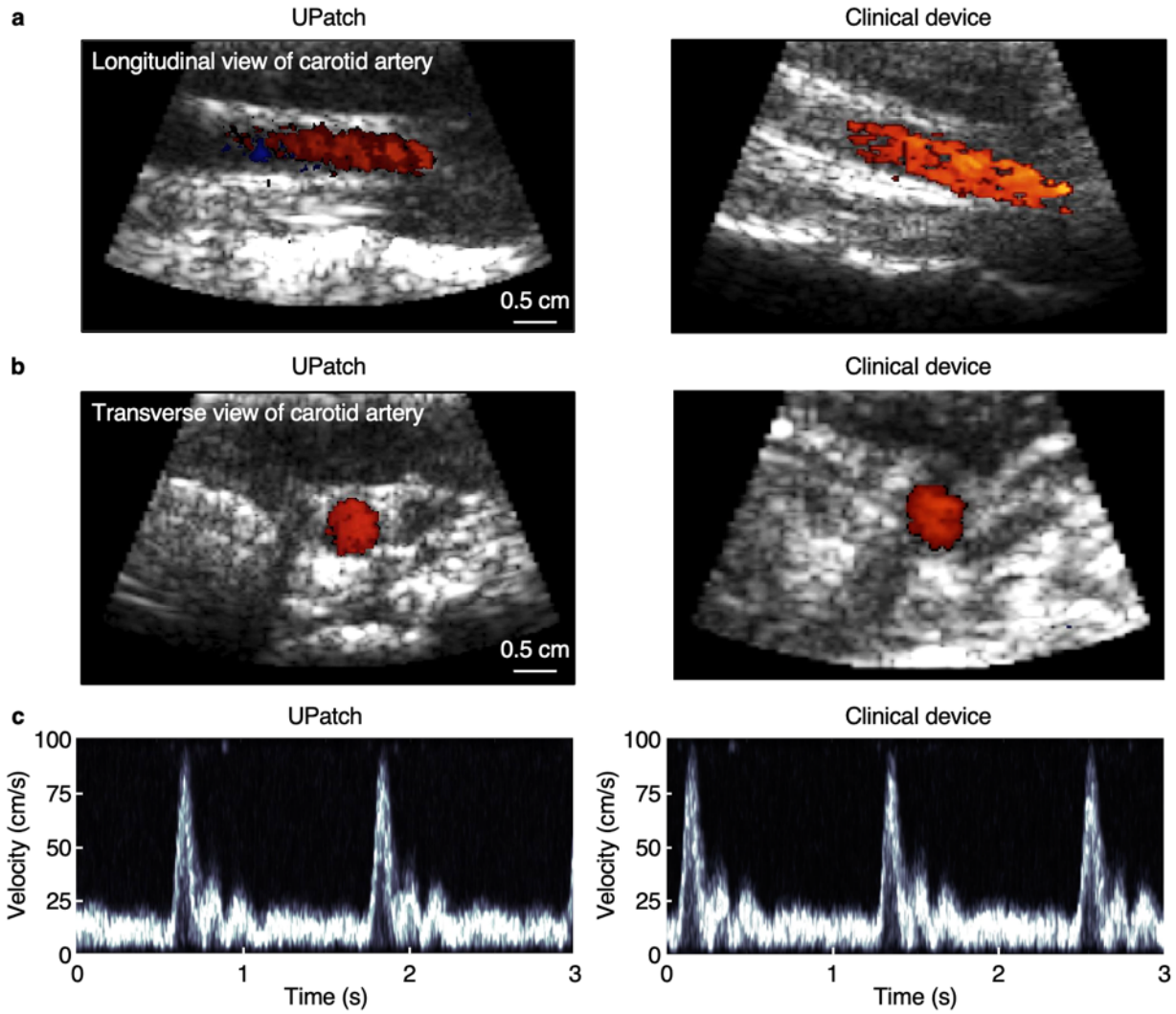
Supplementary Fig. 17 | Blood flow measurement accuracy. a, Mean blood flow velocity at each depth. The black dashed line shows the ground truth mean velocity of 15 cm/s set on the peristaltic pump. **b**, Inaccuracy of the mean blood flow velocity as a function of depth. The dashed line shows the inaccuracy threshold of 10% typically used in ultrasound quality assurance protocols[181]. The two devices demonstrate similar levels of inaccuracies. **c**, Measured mean velocities as a function of different ground truth mean velocities at a depth of 50 mm. The black dashed line represents the linearity between the measured and ground truth velocities. Data in **a** and **c** are presented as mean ± standard deviations (n = 4).



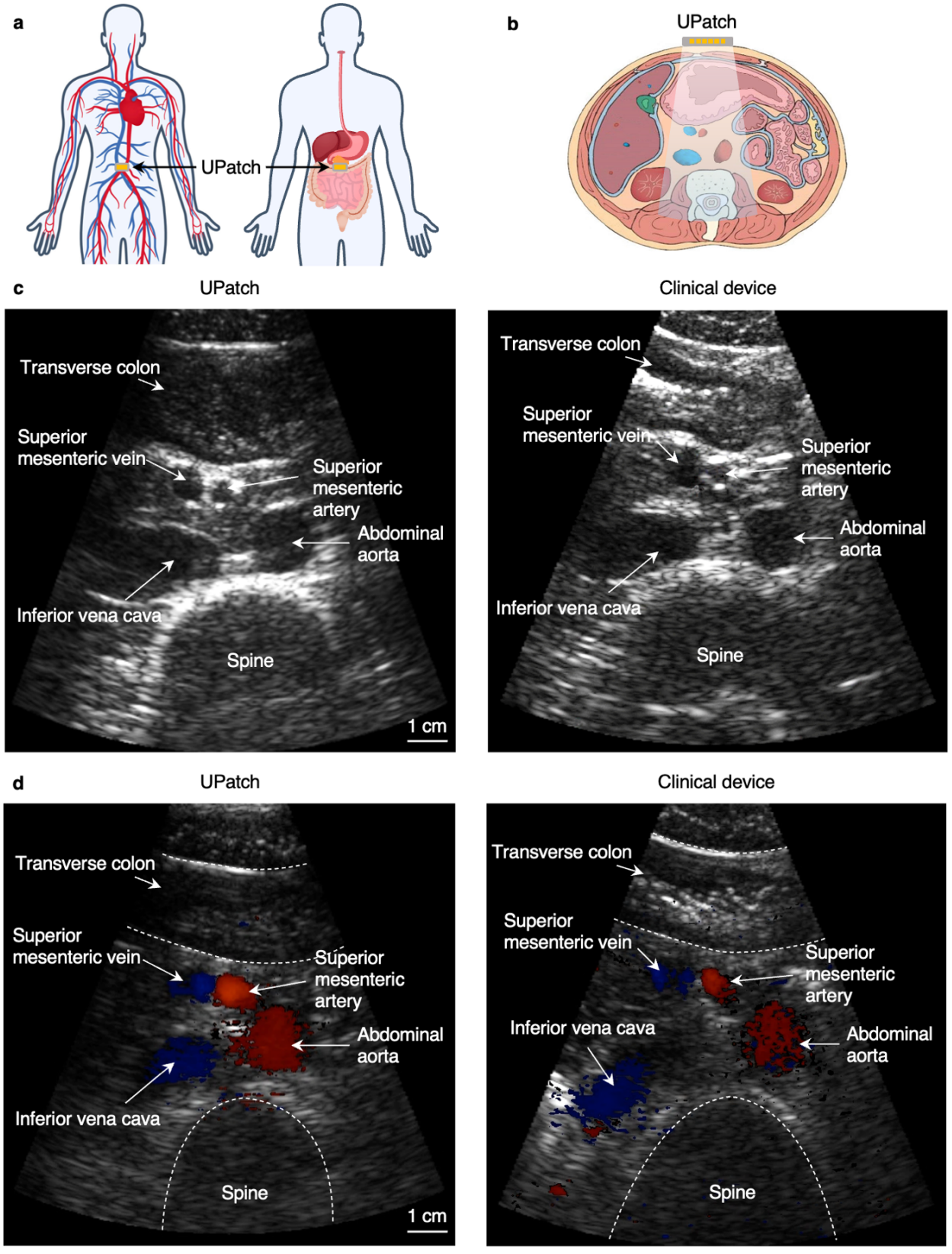
Supplementary Fig. 18 | Design of the umbilical cord phantom. a, Model (top) and image of the stereolithographically printed phantom (bottom) of the helical umbilical cord. The umbilical cord is composed of two arteries and one vein. **b**, Acoustic impedances of different materials. The Elastic 50A resin (Formlabs) exhibits small acoustic impedance mismatch with water and Doppler fluid[85,182], minimizing acoustic reflection at fluid–structure interfaces. In contrast, the greater mismatch of silicone tubing (Metaland) leads to substantial acoustic reflection, particularly when insonating through multiple interfaces of the umbilical cord. Data in **b** are presented as mean ± standard deviations (n = 4).



Supplementary Fig. 19 | Stereolithographic fabrication of the umbilical cord phantom. a, Schematic of the stereolithography process showing layer-by-layer photopolymerization of resin to fabricate an umbilical cord phantom. A vertically translating build platform enables sequential printing within a resin tank. A light source passes through the beam collimator to produce coherent illumination. A dynamic optical mask modulates the illumination to define each layer's geometry. A support structure is printed in parallel with the umbilical cord phantom to ensure geometric stability during stereolithography. **b**, Post-print washing process. The structure is immersed in solvent to remove uncured resin. **c**, Manual removal of support structures to isolate the umbilical cord phantom. **d**, Final polymerization step. An ultraviolet light fully cures the umbilical cord phantom to improve mechanical and chemical robustness.

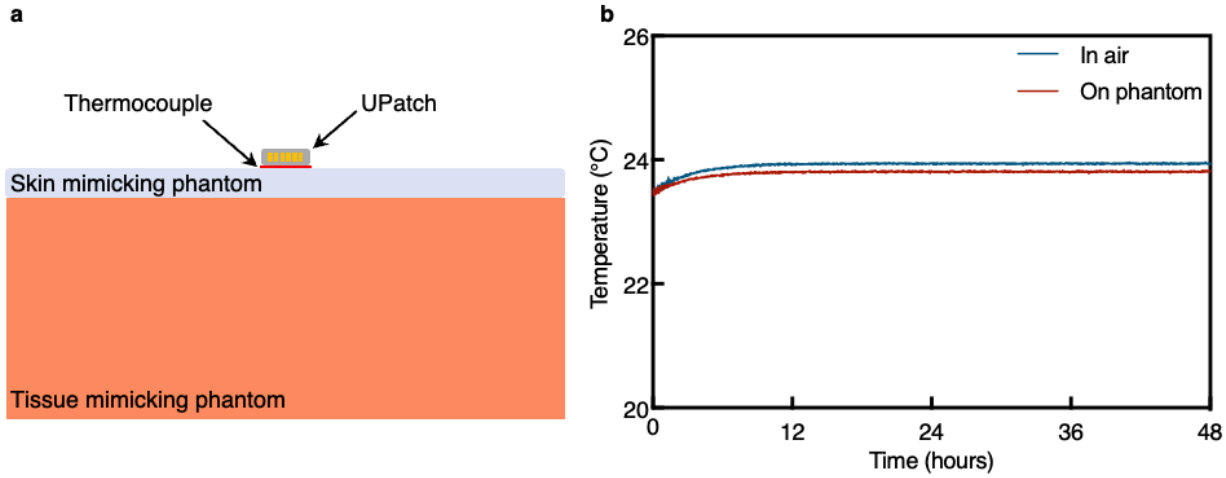


Supplementary Fig. 20 | Measurements on the carotid artery of a healthy adult. Duplex imaging of the carotid artery showing the **a**, longitudinal view and **b**, transverse view with the UPatch (left) and clinical device (right). Anatomy and blood flow of the carotid artery can be recognized from the brightness mode and color Doppler mode, respectively. The images share the same scale bar. **c**, Blood flow measurements of the carotid artery using the spectral Doppler mode. Blood flow spectra obtained from the UPatch (left) and clinical device (right) are comparable on the same participant.

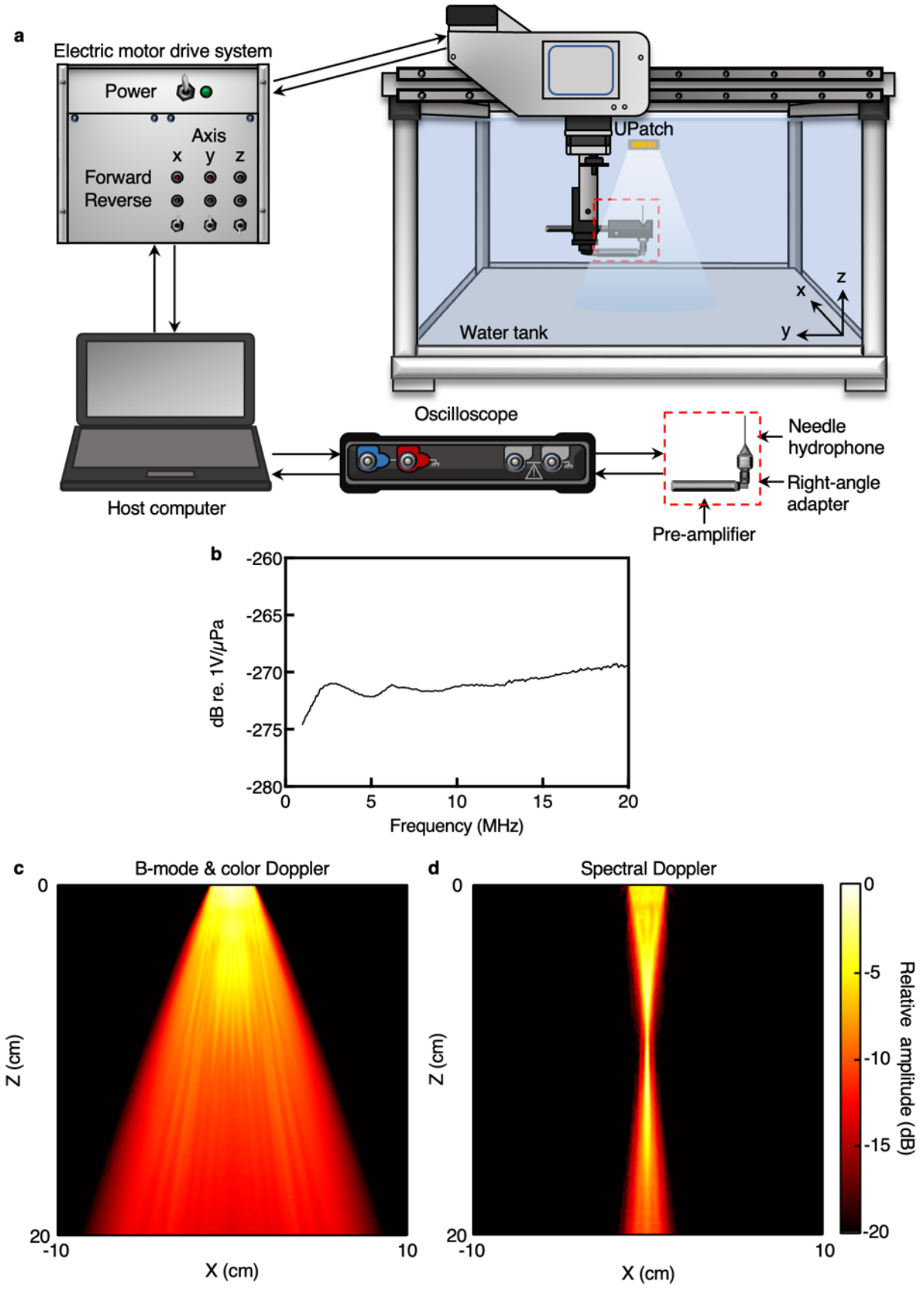


Supplementary Fig. 21 | Measurements on the abdominal cross-section of a healthy adult. a, Schematics showing the position of the UPatch imaging the cardiovascular system (left) and

digestive system (right). **b**, Schematics illustrating the transverse anatomical view of the abdomen. Transverse view of the abdominal anatomy in **c**, brightness and **d**, duplex modes. Both the UPatch and clinical device can clearly recognize the structures of the transverse colon, superior mesenteric vein, superior mesenteric artery, inferior vena cava, abdominal aorta, and spine. The images share the same scale bar.

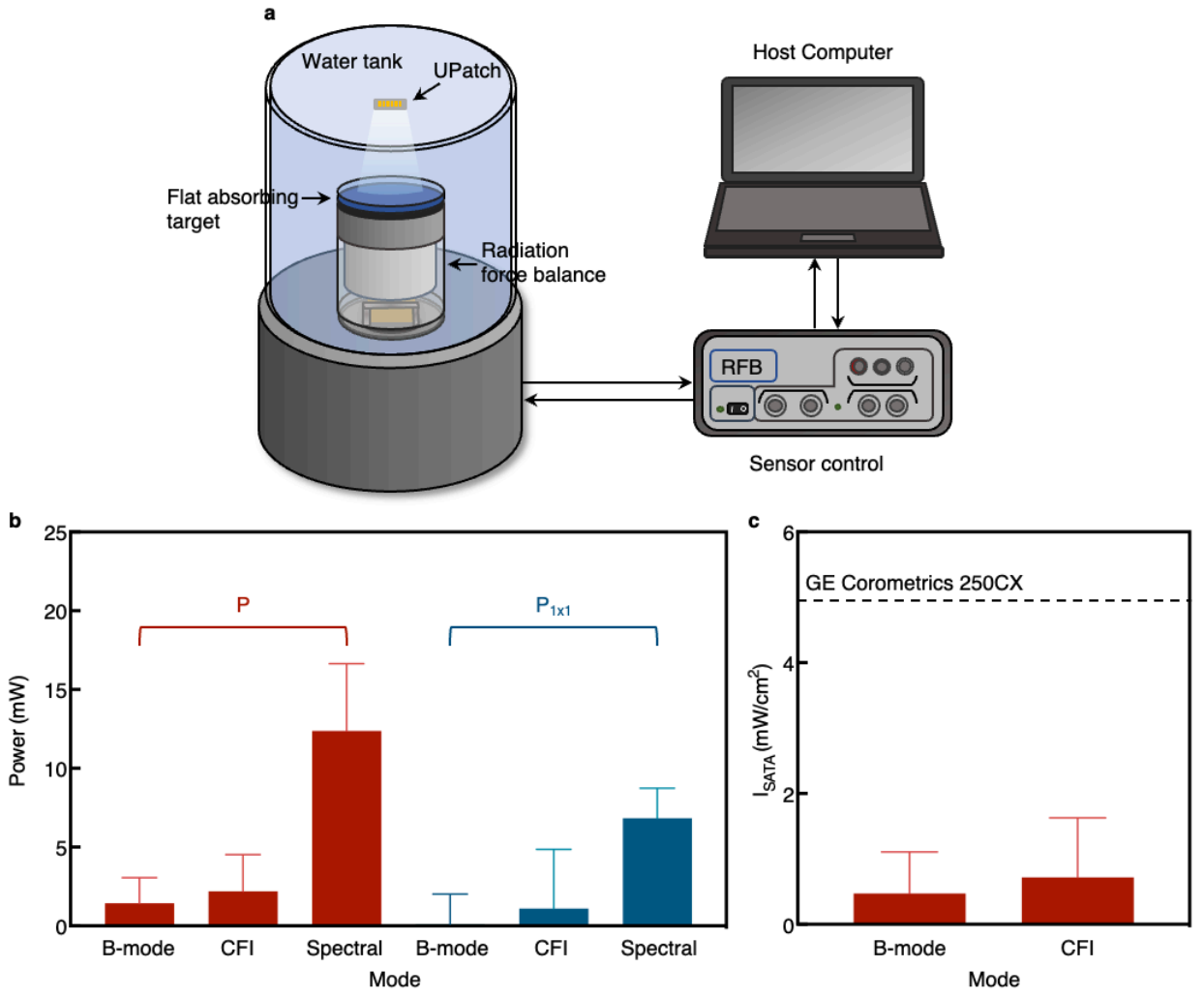


Supplementary Fig. 22 | Surface temperature characterizations. a, Schematic setup for the on phantom testing. The setup includes a skin and tissue mimic phantom (NPL, UK) with similar acoustic and thermal properties to those of human skin and tissues. The thermocouple was placed in between the UPatch and the skin mimic phantom during the measurement. **b**, Surface temperature measurements on phantom and in air. The in-air test was performed by placing the UPatch and thermocouple in still air[183]. The maximum temperature increase was ~ 0.53 °C, which was < 0.7 °C for continuous monitoring recommended by the AIUM[123] and BMUS[124].

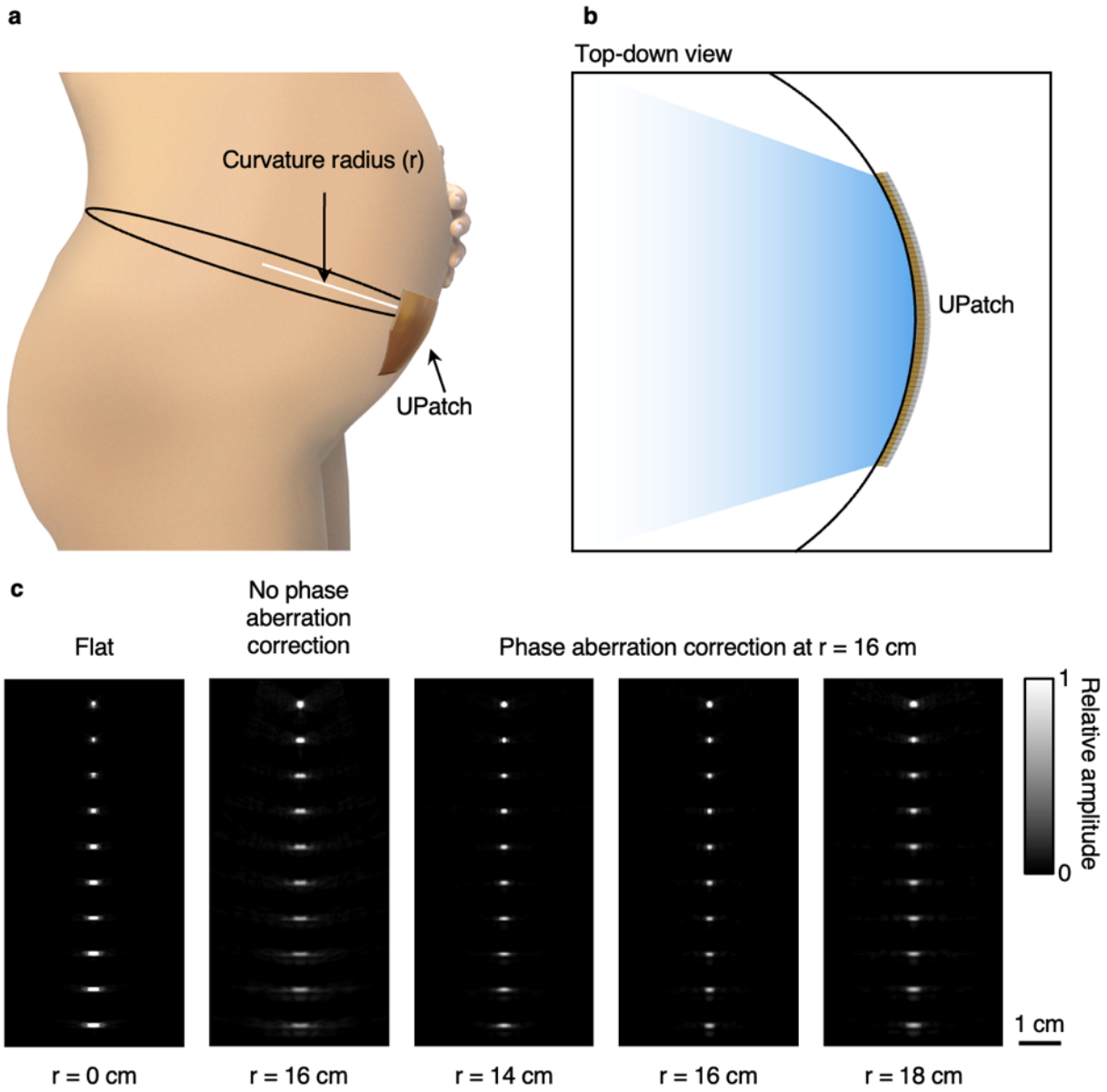

a
Electric motor drive system
Power
Axis
x y z
Forward
Reverse
UPatch
Water tank
z
x
y
Oscilloscope
Needle hydrophone
Right-angle adapter
Pre-amplifier
Host computer
b
dB re. 1V/μPa
-260
-265
-270
-275
-280
0
5
10
15
20
Frequency (MHz)
c
B-mode & color Doppler
0
20
-10
10
Z (cm)
X (cm)
d
Spectral Doppler
0
20
-10
10
Z (cm)
X (cm)
0
-5
-10
-15
-20
Relative amplitude (dB)

1465

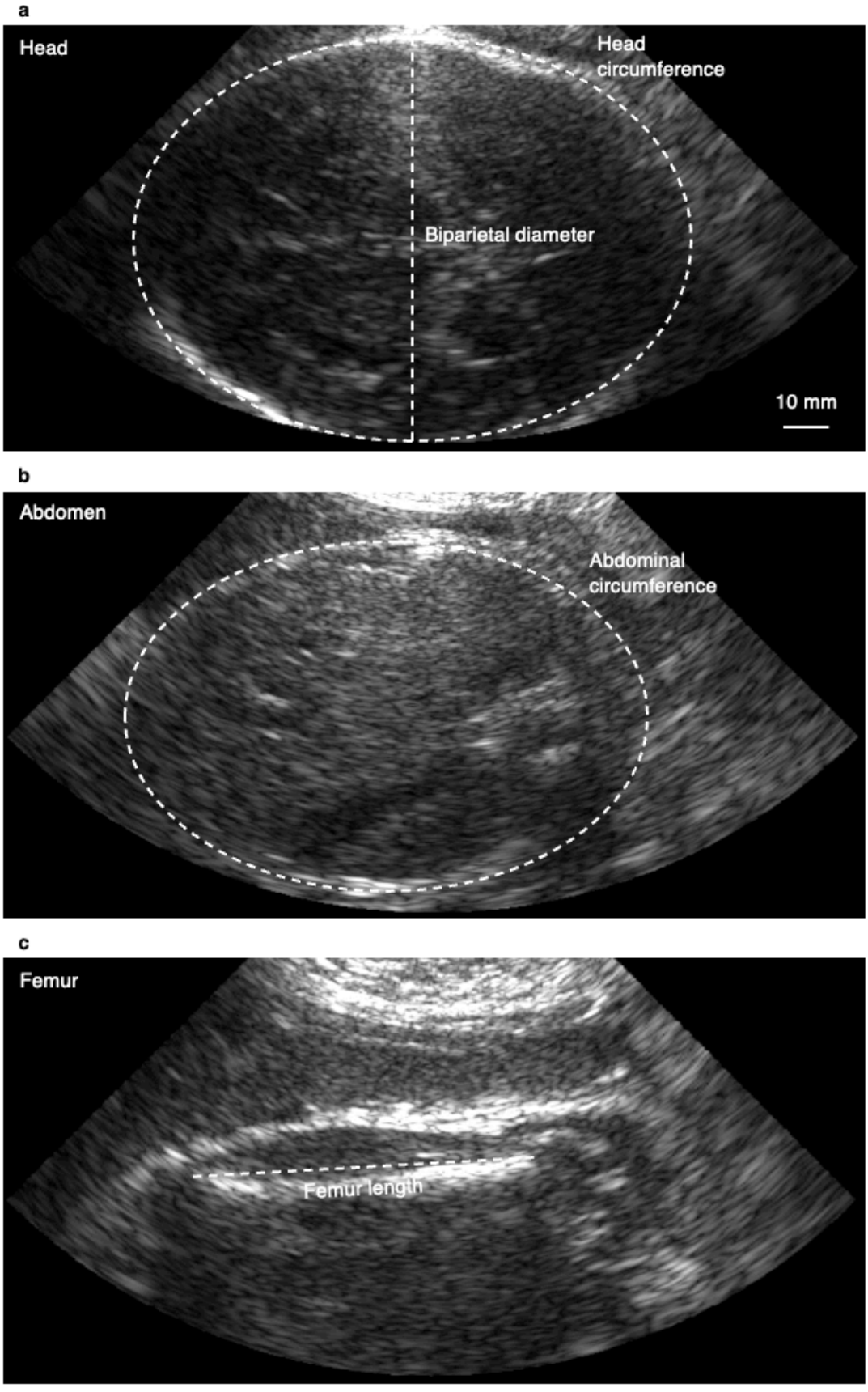
Supplementary Fig. 23 | Safety measurements with a hydrophone. a, Schematic setup with a hydrophone system. Ultrasound intensity from the UPatch can be measured in a water tank. **b**, Magnitude data from calibrating the hydrophone sensitivity for accurate calculation of its acoustic properties. Acoustic field scans of **c**, a diverging beam (for brightness and color-flow modes) and **d**, a focused beam (for the spectral Doppler mode) with the UPatch in the XZ plane. The images share the same color bar.



Supplementary Fig. 24 | Safety measurements with a radiation force balance. a, Schematic setup with a radiation force balance. The ultrasound beam from the UPatch is uniformly absorbed in the flat absorbing target and the radiation force balance can accurately measure the power of the transmitted beam. **b**, Power (P) and bounded-square output power over a one square centimeter area (P_{1x1}) were measured in brightness (B-mode), color-flow imaging (CFI), and spectral Doppler modes. **c**, Spatial average temporal average intensity (I_{SATA}) derived from power (Supplementary Discussion 11). The black dashed line represents I_{SATA} of GE Corometrics 250CX fetal heart rate monitors[18]. Data in **b** and **c** are presented as mean ± standard deviations (n = 4).

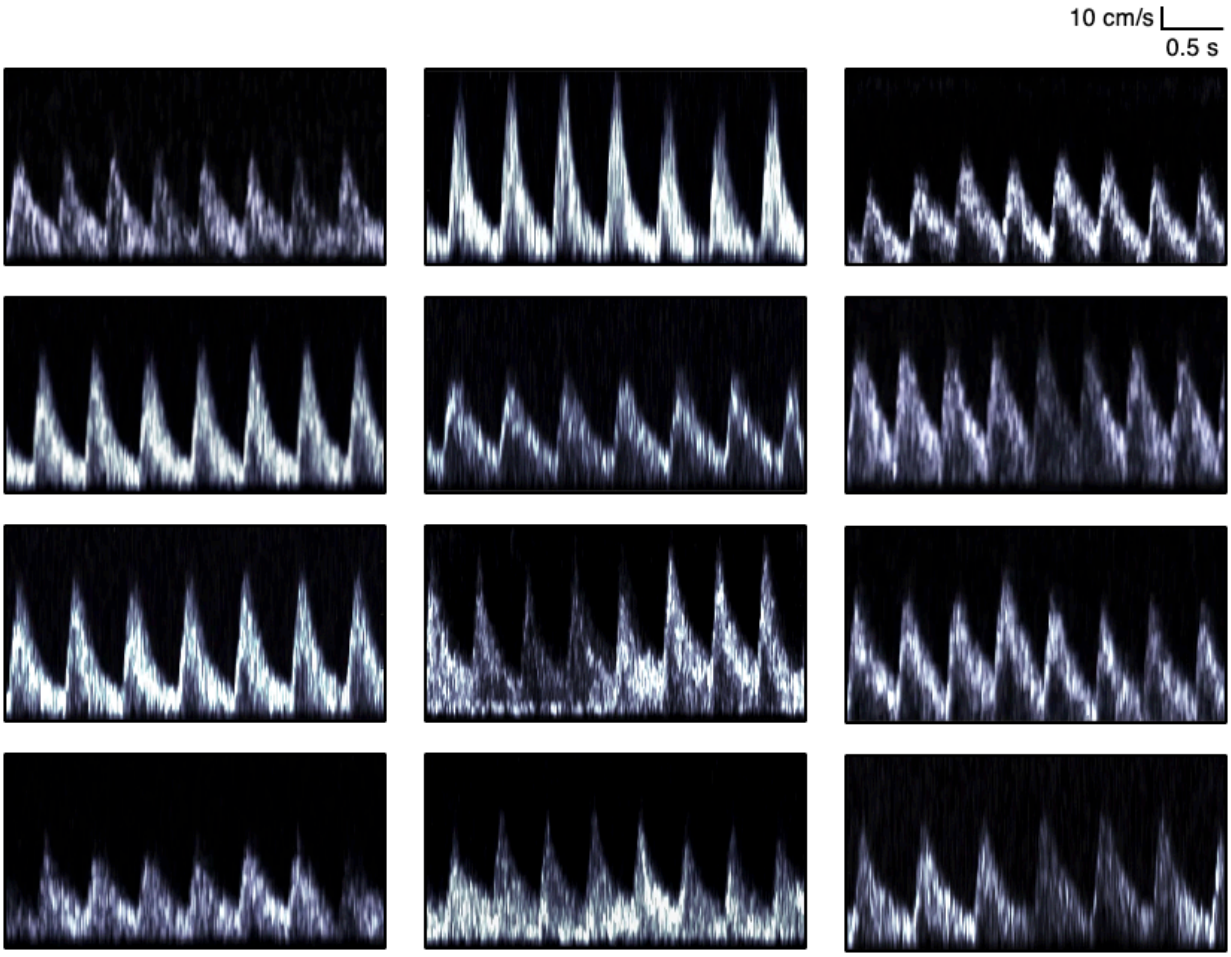


Supplementary Fig. 25 | Deformation of the UPatch. a, Schematics of the UPatch on the maternal abdomen. **b**, Top-down view of the UPatch on the maternal abdomen. **c**, Simulation results of the image quality. The maternal abdominal curvatures can induce phase aberration and affect the image quality of the UPatch. To address this, the phase aberration was corrected at a curvature radius of 16 cm, which encompasses the typical curvature radii from 14 cm to 18 cm in the third trimester[128,129]. The phase aberration correction improved the image quality within this curvature range. Simulations were performed with the Matlab Field II toolbox. The images share the same scale bar.

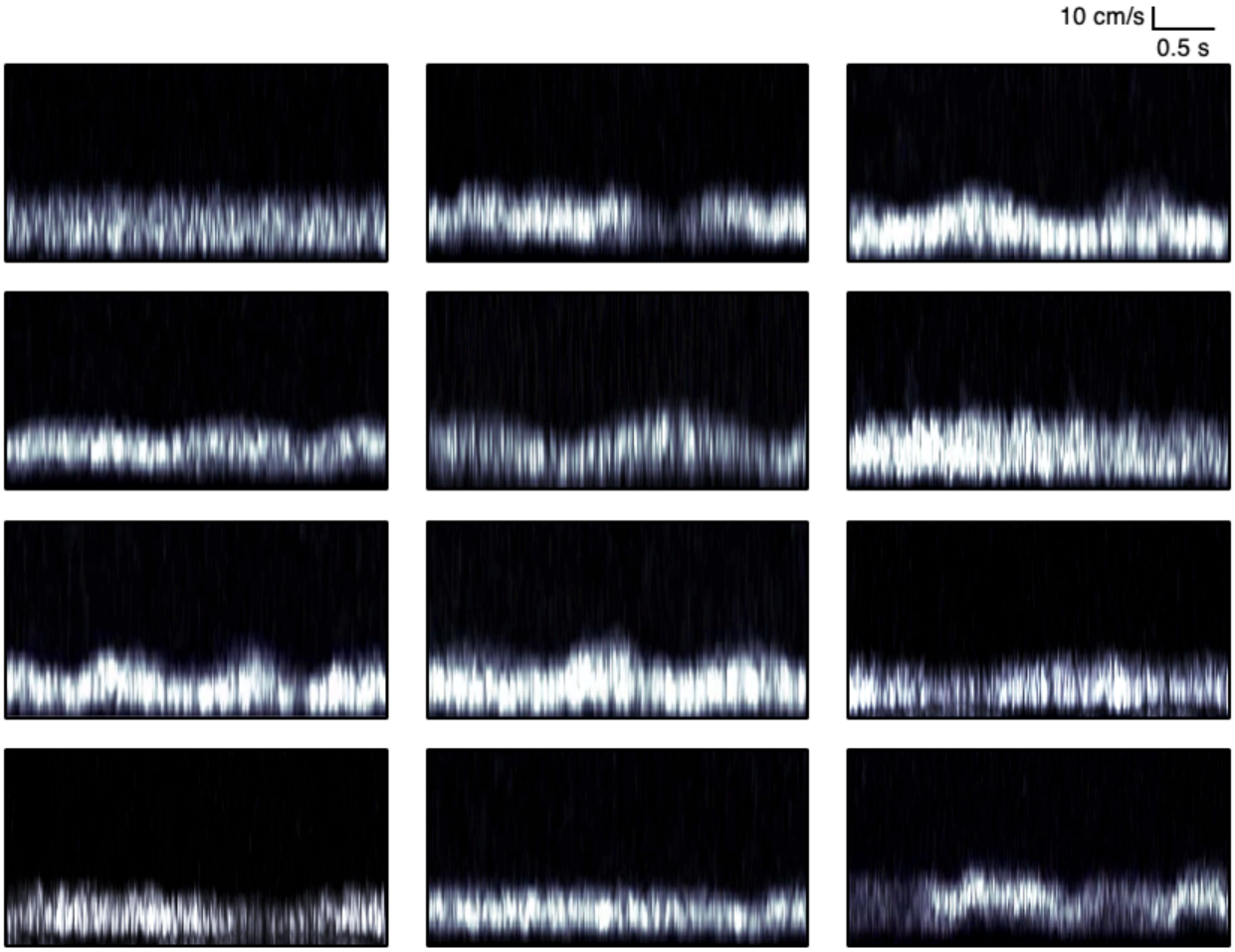

a
Head
Head
circumference
Biparietal diameter
10 mm
b
Abdomen
Abdominal
circumference
c
Femur
Femur length

1490

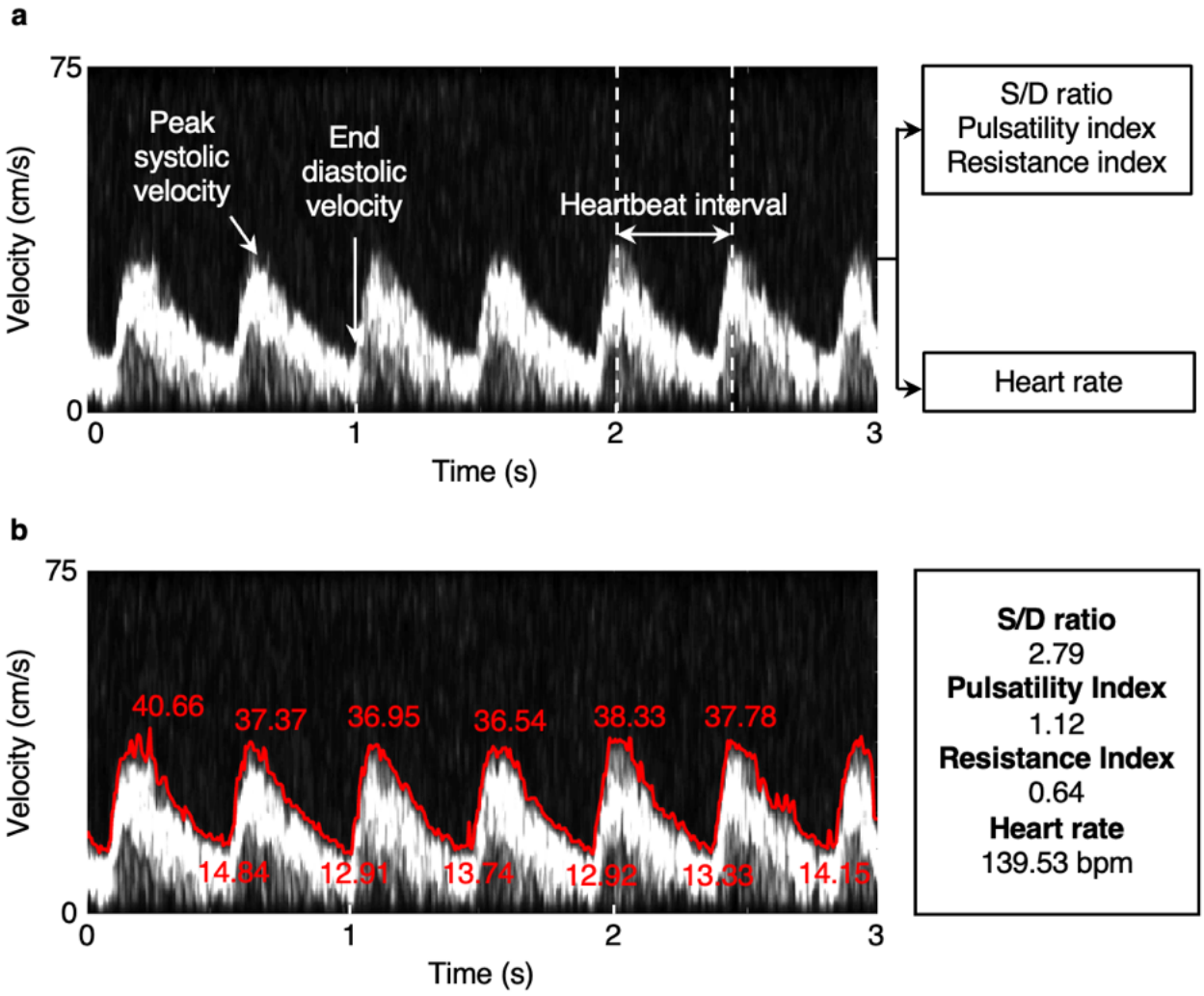
Supplementary Fig. 26 | Fetal biometry. Brightness-mode images of a fetal **a**, head, **b**, abdomen, and **c**, femur. All images share the same scale bar. Biparietal diameter, head circumference, abdominal circumference, and femur length can be measured to calculate the estimated fetal weight using the Hadlock IV formula[184]. For this particular fetus, its weight is calculated to be 3.56 kg using the UPatch and 3.54 kg using the clinical device.



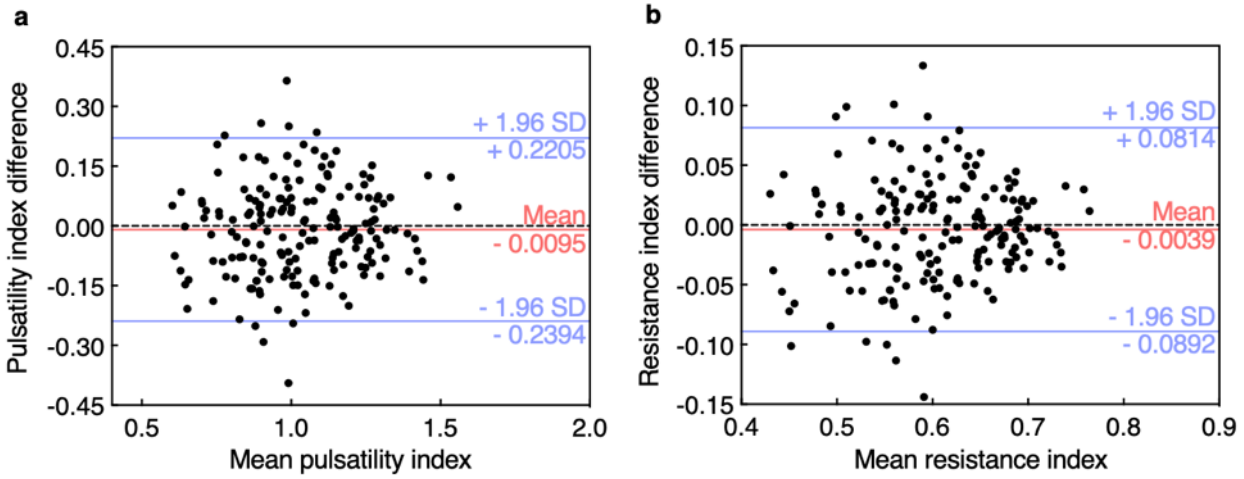
Supplementary Fig. 27 | Spectral Doppler of the umbilical artery. The UPatch can acquire spectral Doppler signals from the umbilical artery of pregnant participants. All signals were acquired by a sonographer specializing in fetal ultrasonography. All images share the same scale bars. Each panel corresponds to the same participant and ordering as in Extended Data Fig. 3.



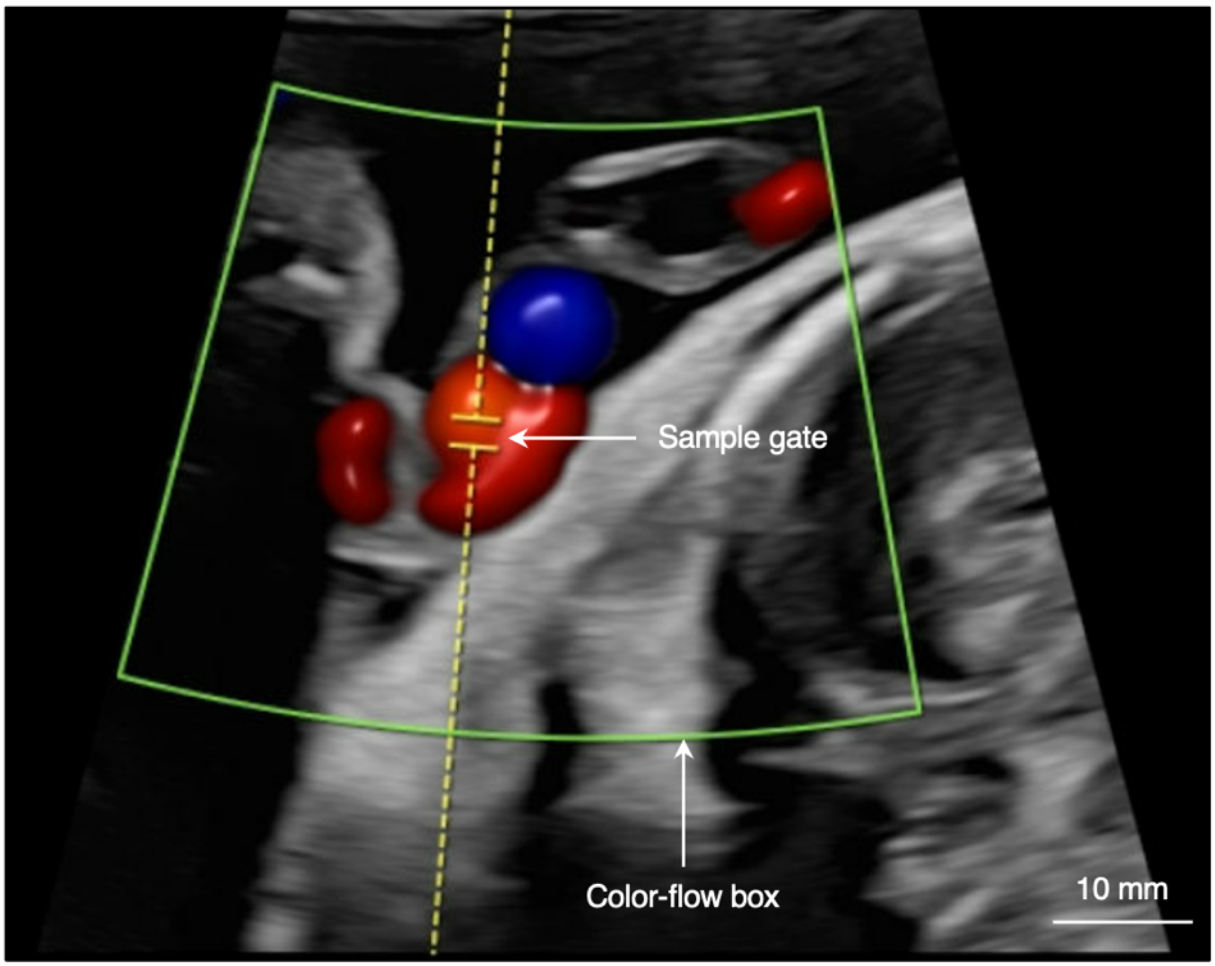
Supplementary Fig. 28 | Spectral Doppler of the umbilical vein. The UPatch can acquire spectral Doppler signals from the umbilical vein of pregnant participants. All signals were acquired by a sonographer specializing in fetal ultrasonography. All images share the same scale bars. Each panel corresponds to the same participant and ordering as in Extended Data Fig. 3 and Supplementary Fig. 27.



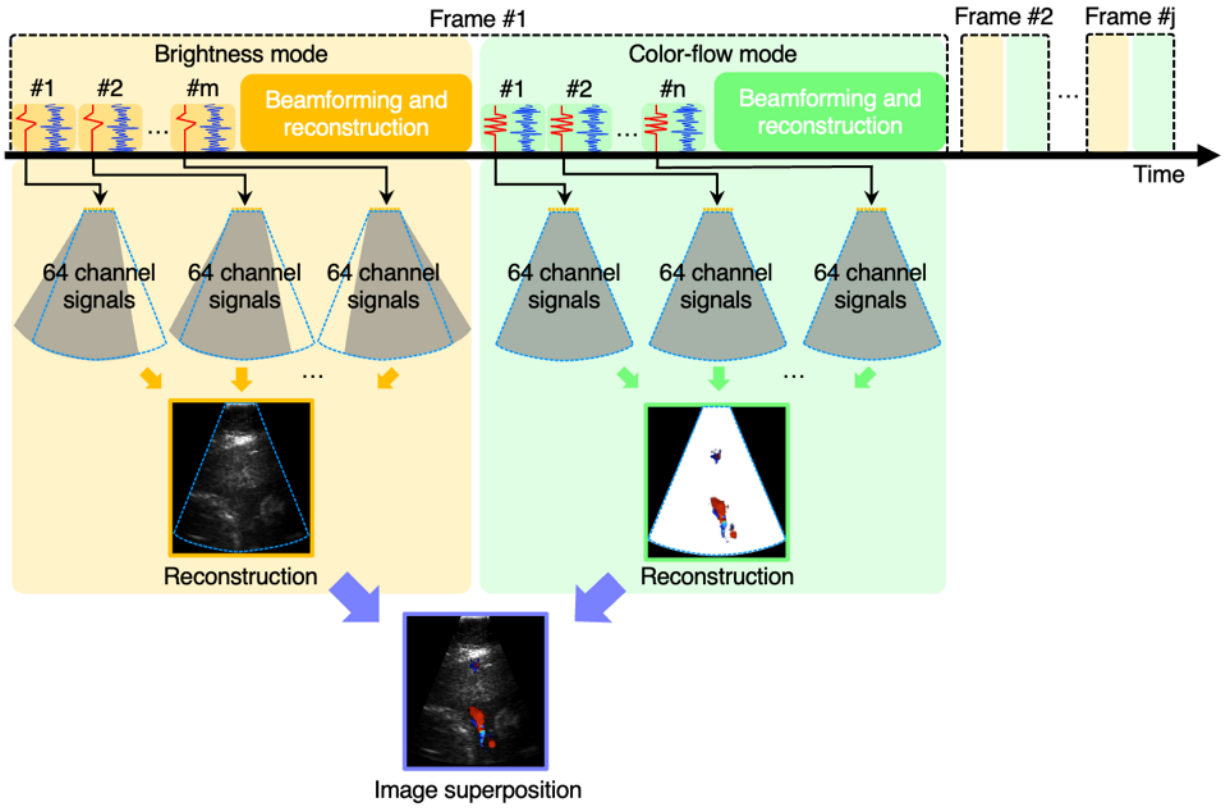
Supplementary Fig. 29 | Envelope extraction from spectral Doppler signals. a, Spectral Doppler signals from the fetal umbilical artery. Peak systolic velocity, end diastolic velocity, and heartbeat intervals can be manually identified and used to derive the systolic-to-diastolic (S/D) ratio, pulsatility index, resistance index, and heart rate. **b**, The envelope extraction algorithm we developed in this study allows automated derivation of these parameters[73].



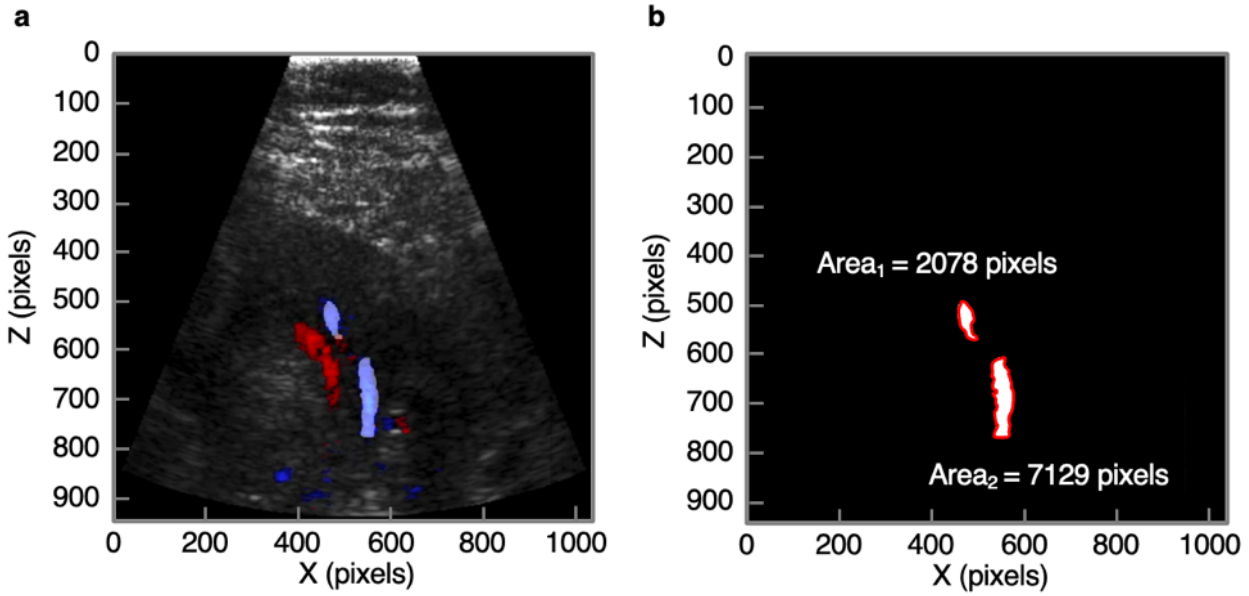
Supplementary Fig. 30 | Statistical analysis of the pulsatility index and the resistance index. Bland-Altman plots of **a**, the pulsatility index and **b**, the resistance index measured using the UPatch and a handheld clinical ultrasound device (Voluson E10, GE). Solid red lines are the mean differences between the two devices, solid blue lines are 95% limits of agreement (1.96 standard deviations above and below the mean differences), and black dashed lines are the zero difference between the two devices. Each plot has three measurement pairs by repeating three times with the same devices on the same participant for each of the 62 participants.



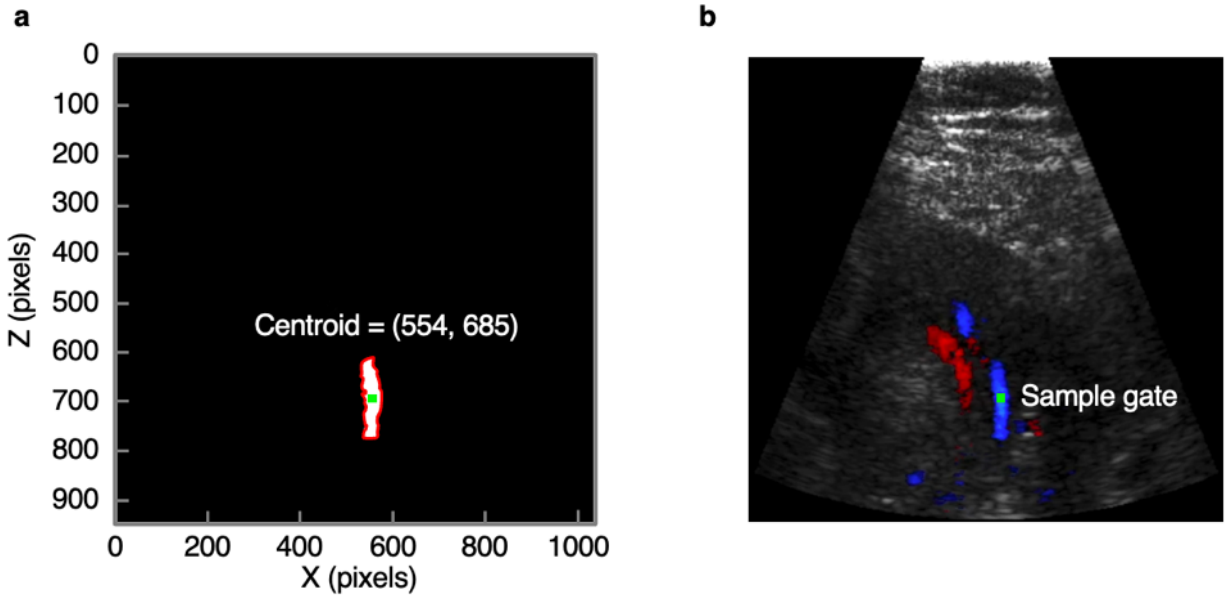
Supplementary Fig. 31 | Imaging with a handheld clinical ultrasound device. Duplex imaging of the umbilical cord with a handheld clinical ultrasound device (GE Voluson E10) requires maneuvering by a sonographer. First, a color-flow box (green) is manually moved to find the color-flow signals from the umbilical cord. Then, a sample gate (yellow) is chosen inside the color-flow box for spectral Doppler recording.



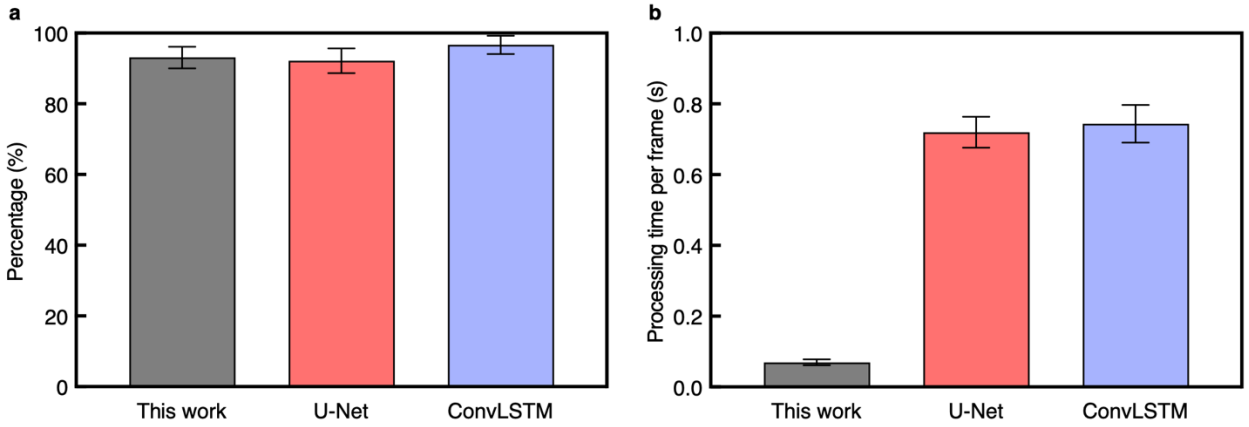
Supplementary Fig. 32 | Data acquisition processes of the UPatch. The UPatch requires a complex beamforming process to synthesize an image from 64 channels. Reconstructing a brightness-mode image requires compounding radiofrequency signals from multiple steered beams. Reconstructing a color-flow image requires estimating the mean frequency across multiple transmitted and received beams. These two images are overlaid to form a duplex image.



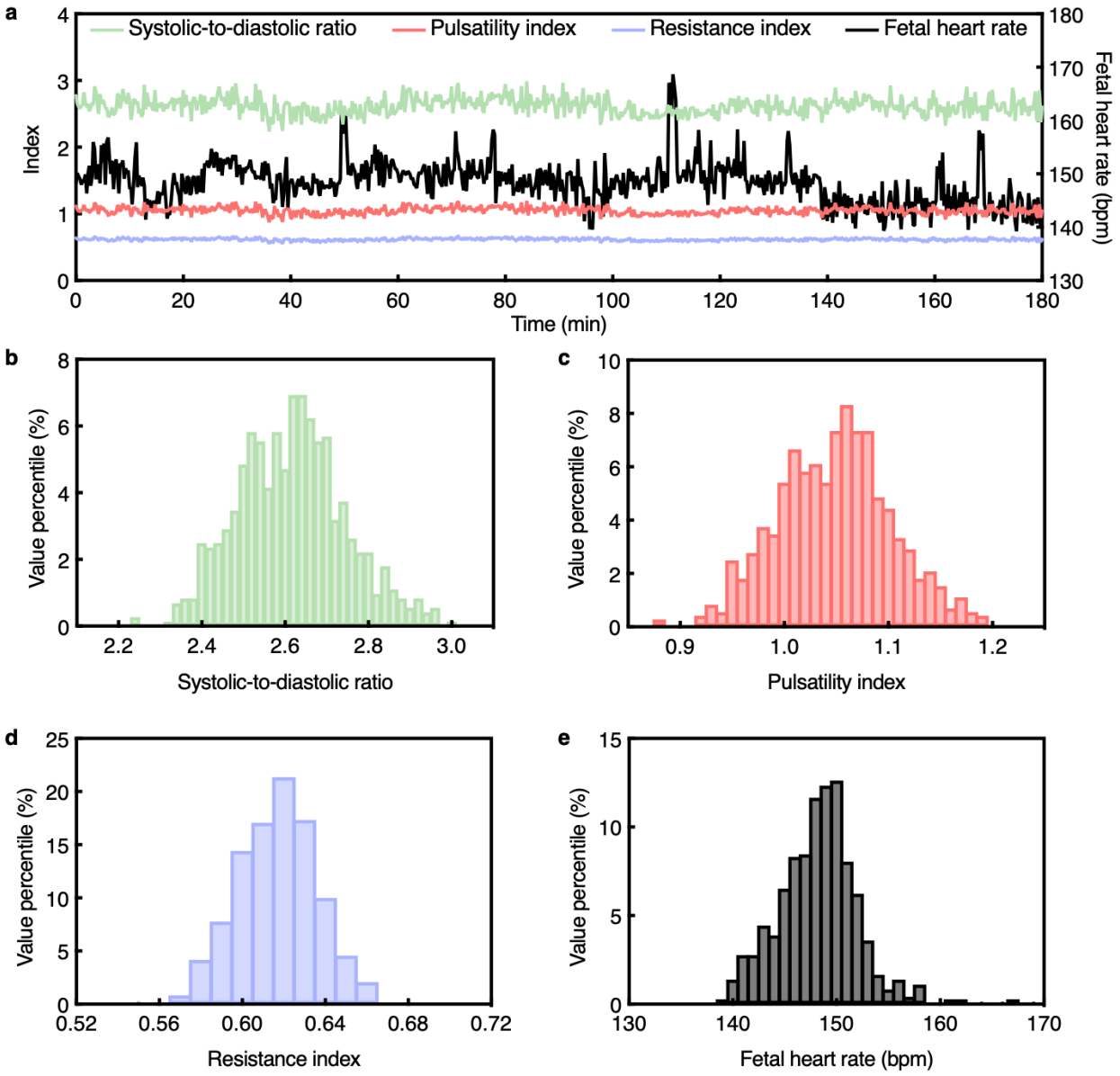
Supplementary Fig. 33 | Recognition of the primary region of the umbilical artery. a, Duplex image with assigned pixel coordinates in the XZ plane for vessel localization. **b**, Number of pixels for each segmented arterial region. The region with the largest area ($Area_2$ in this case) was designated as the primary region.



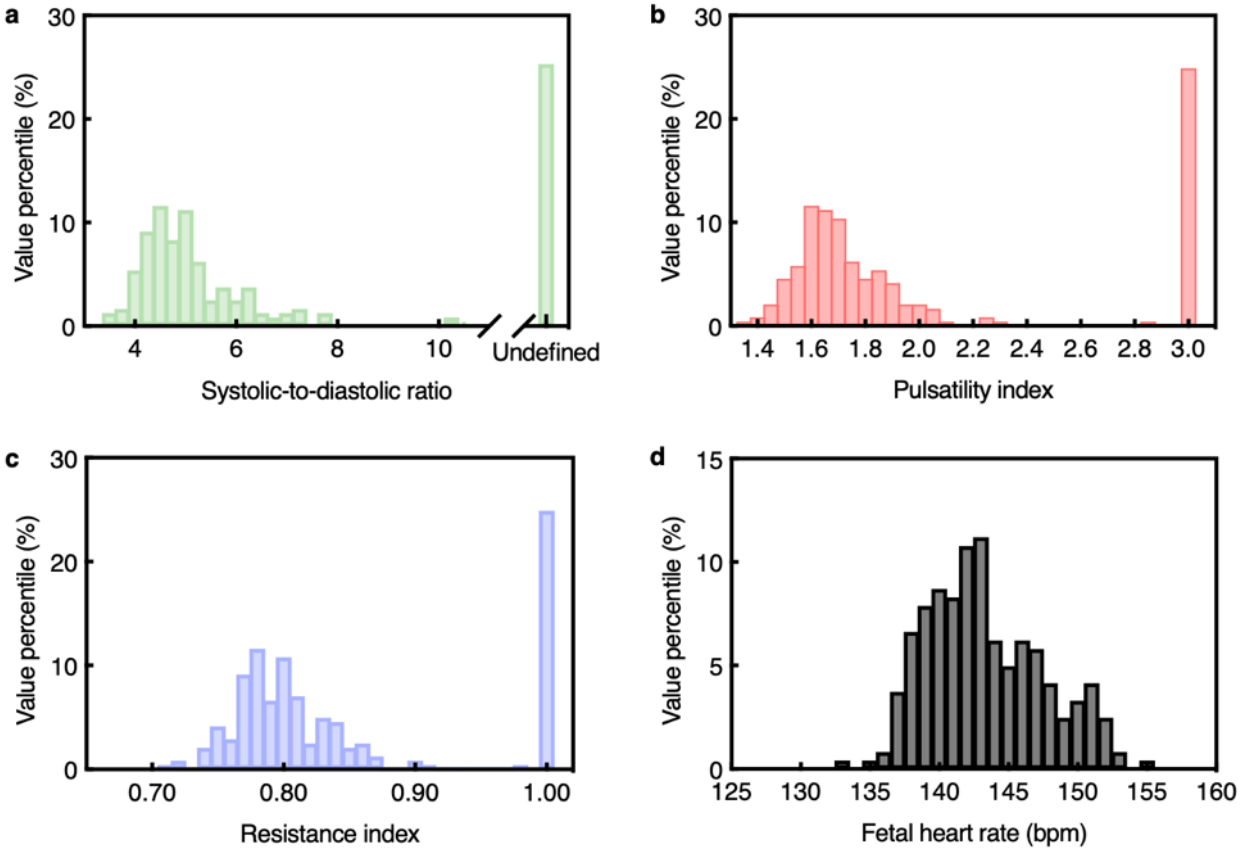
Supplementary Fig. 34 | Identification of the centroid for sample gate placement. a, Centroid calculation of the segmented primary arterial region. The centroid was computed using the geometric center of the pixel distribution. **b**, The sample gate was placed at the calculated centroid position within the duplex image. The color of the artery in the duplex image, whether toward (warm hue) or away from (cool hue) the UPatch, does not affect the centroid placement.



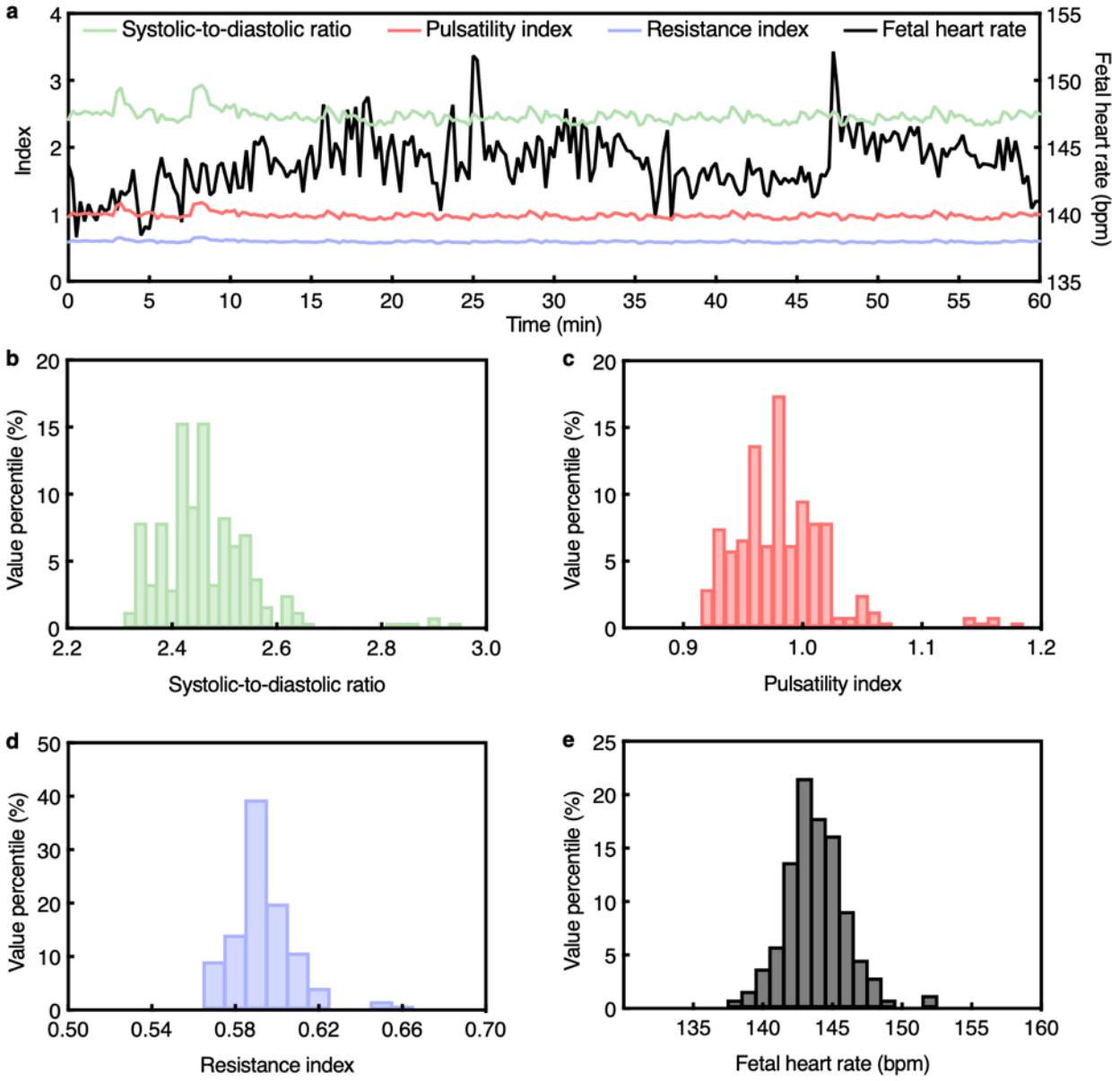
Supplementary Fig. 35 | Benchmarking the tracking algorithm against deep learning methods. a, Sonographer-evaluated tracking accuracy for the algorithm ("This work"), U-Net model, and ConvLSTM model, reported as "optimal" by the sonographers. **b**, Processing time per frame for each method. Data are presented as mean ± standard deviations (n = 3 for **a** and n = 50 for **b**). While all three approaches achieve comparable tracking accuracy, the customized algorithm operates at substantially lower computational cost, enabling real-time tracking during fetal movement.



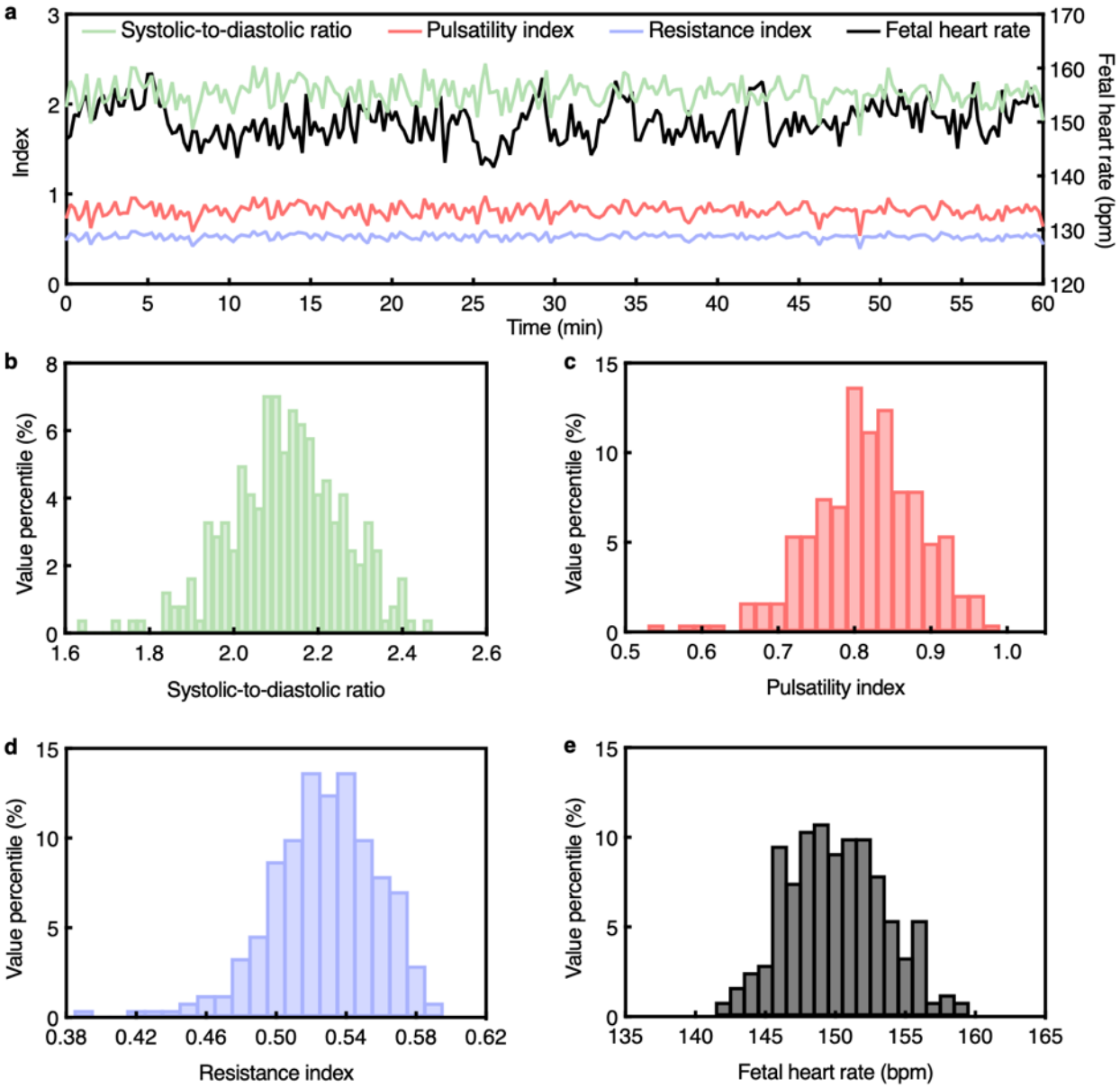
Supplementary Fig. 36 | Continuous fetal monitoring using the UPatch in Participant #1. a, Time series data from the 3-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



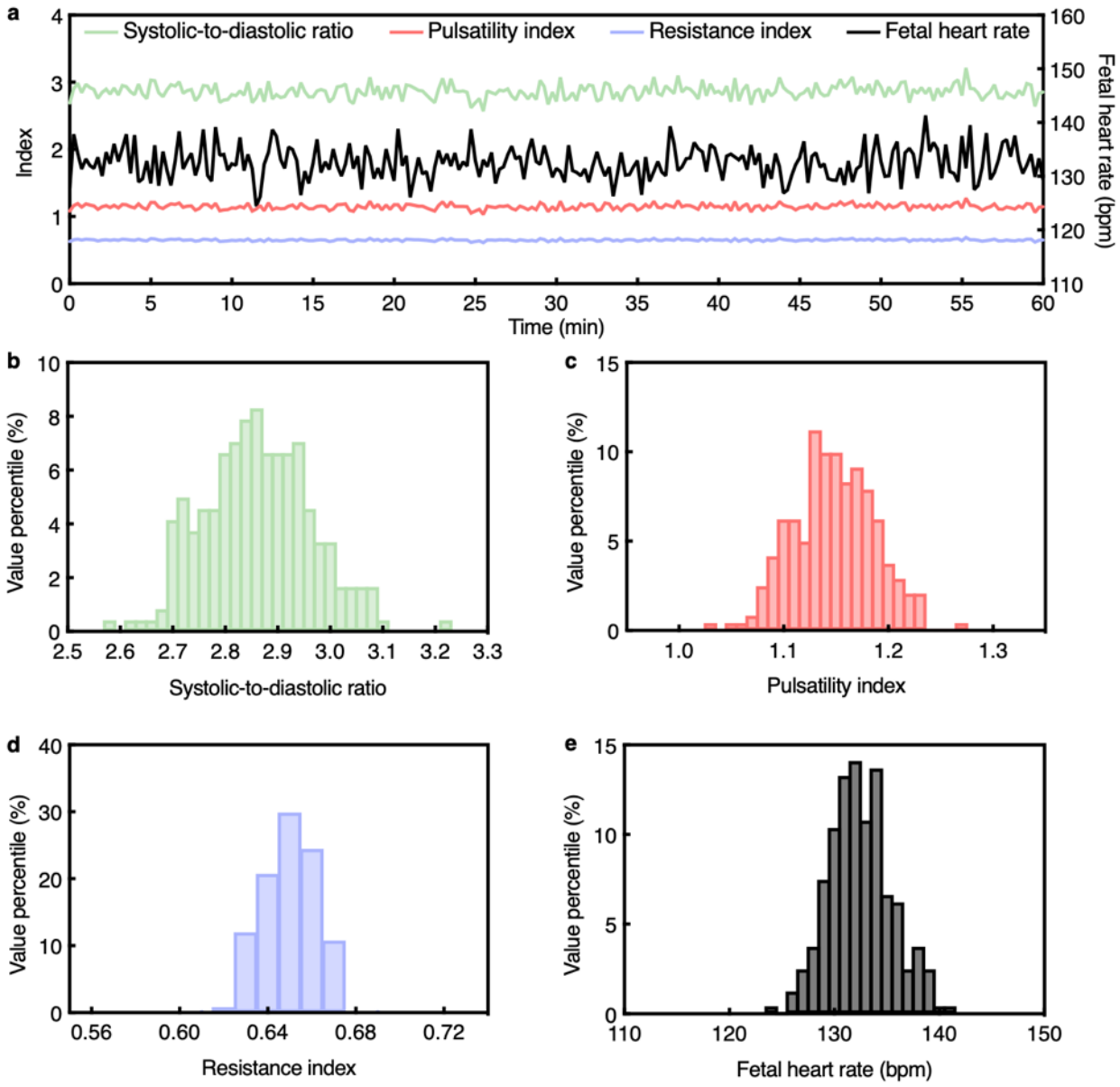
Supplementary Fig. 37 | Continuous fetal monitoring using the UPatch in Participant #2. Histograms of **a**, systolic-to-diastolic ratio, **b**, pulsatility index, **c**, resistance index, and **d**, fetal heart rate over the recording period. The time series data is in Fig. 4d in the main text. The systolic-to-diastolic ratio was not plotted in the time series due to the presence of undefined values.



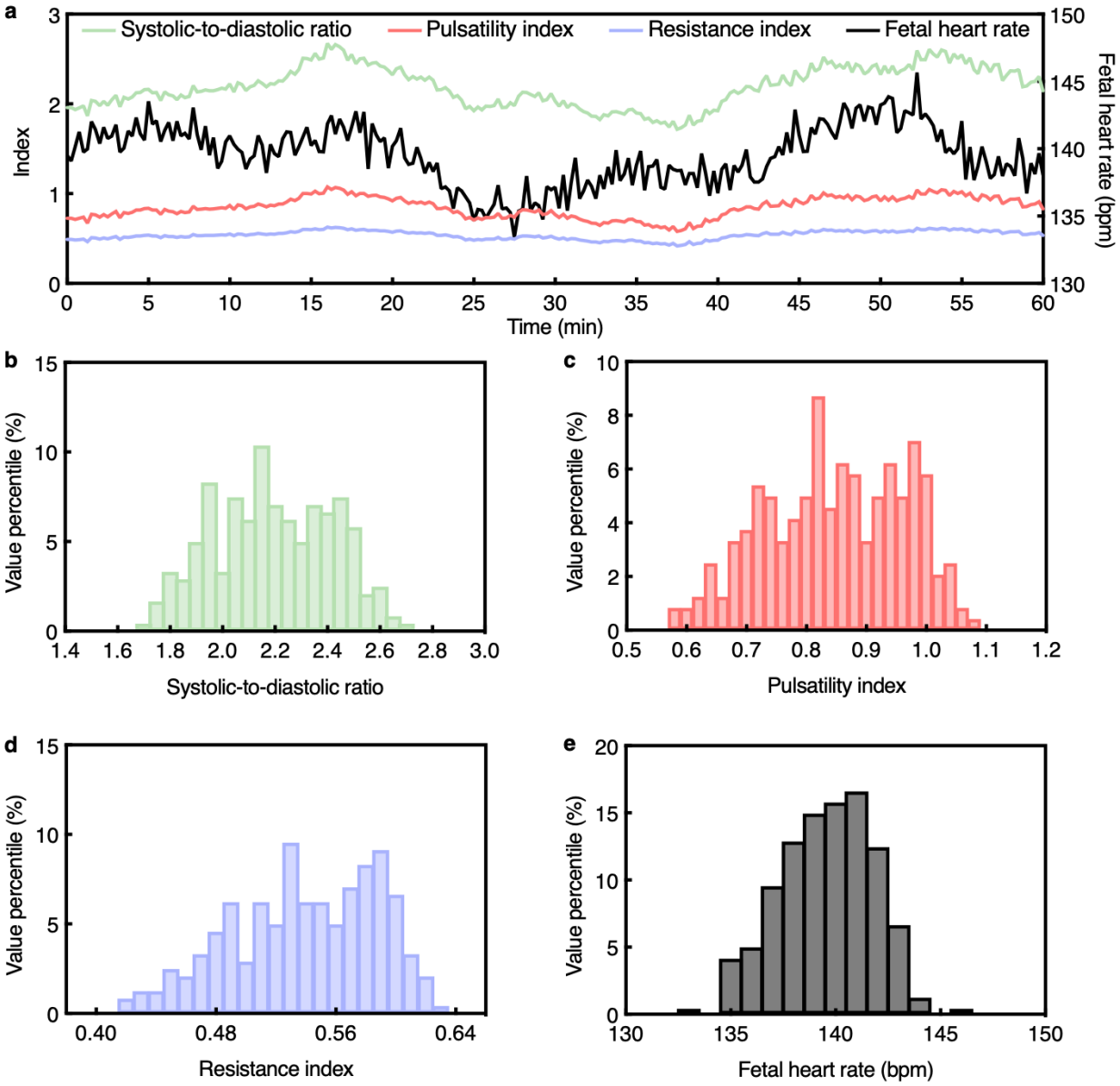
Supplementary Fig. 38 | Continuous fetal monitoring using the UPatch in Participant #3. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



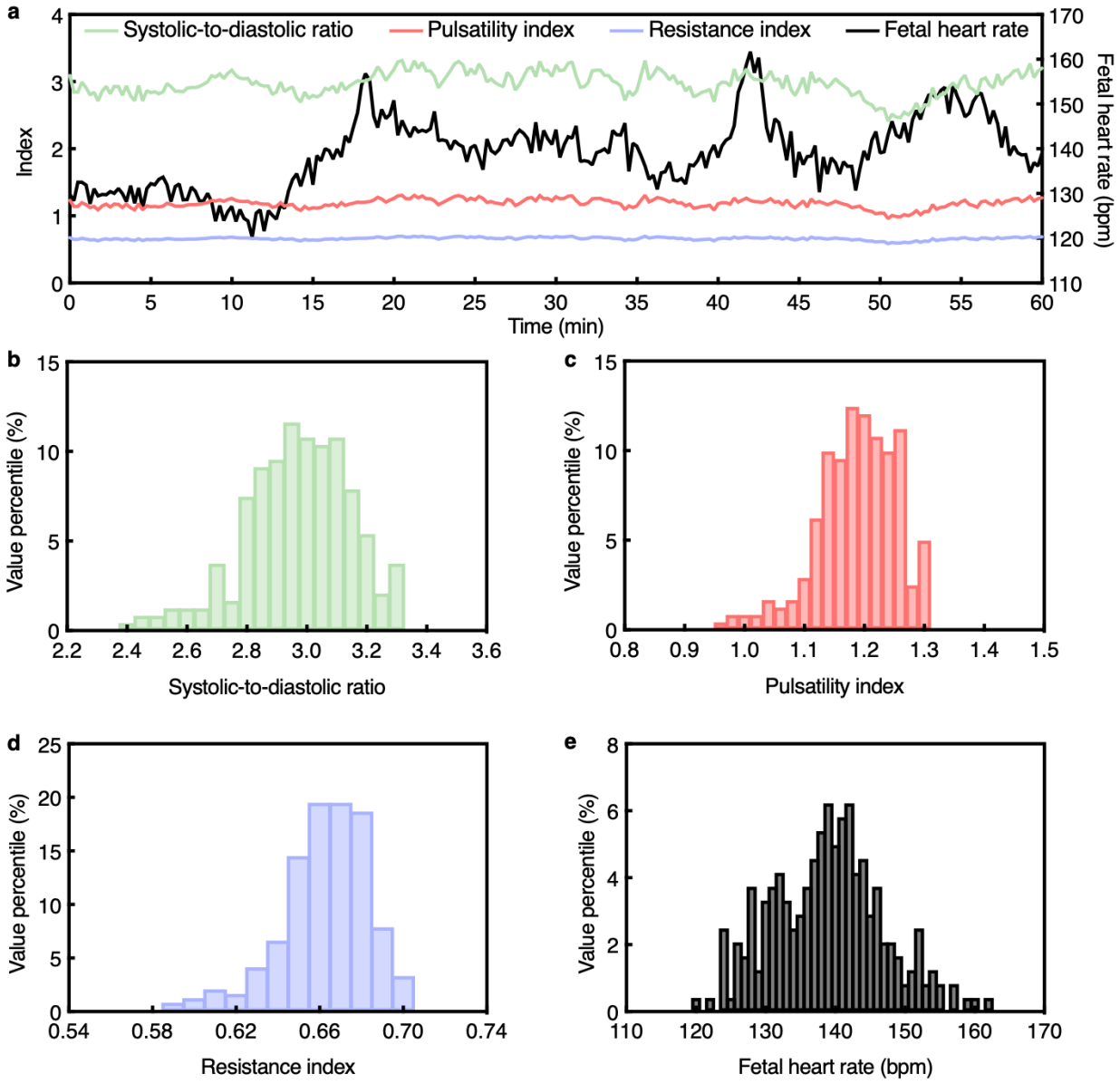
Supplementary Fig. 39 | Continuous fetal monitoring using the UPatch in Participant #4. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



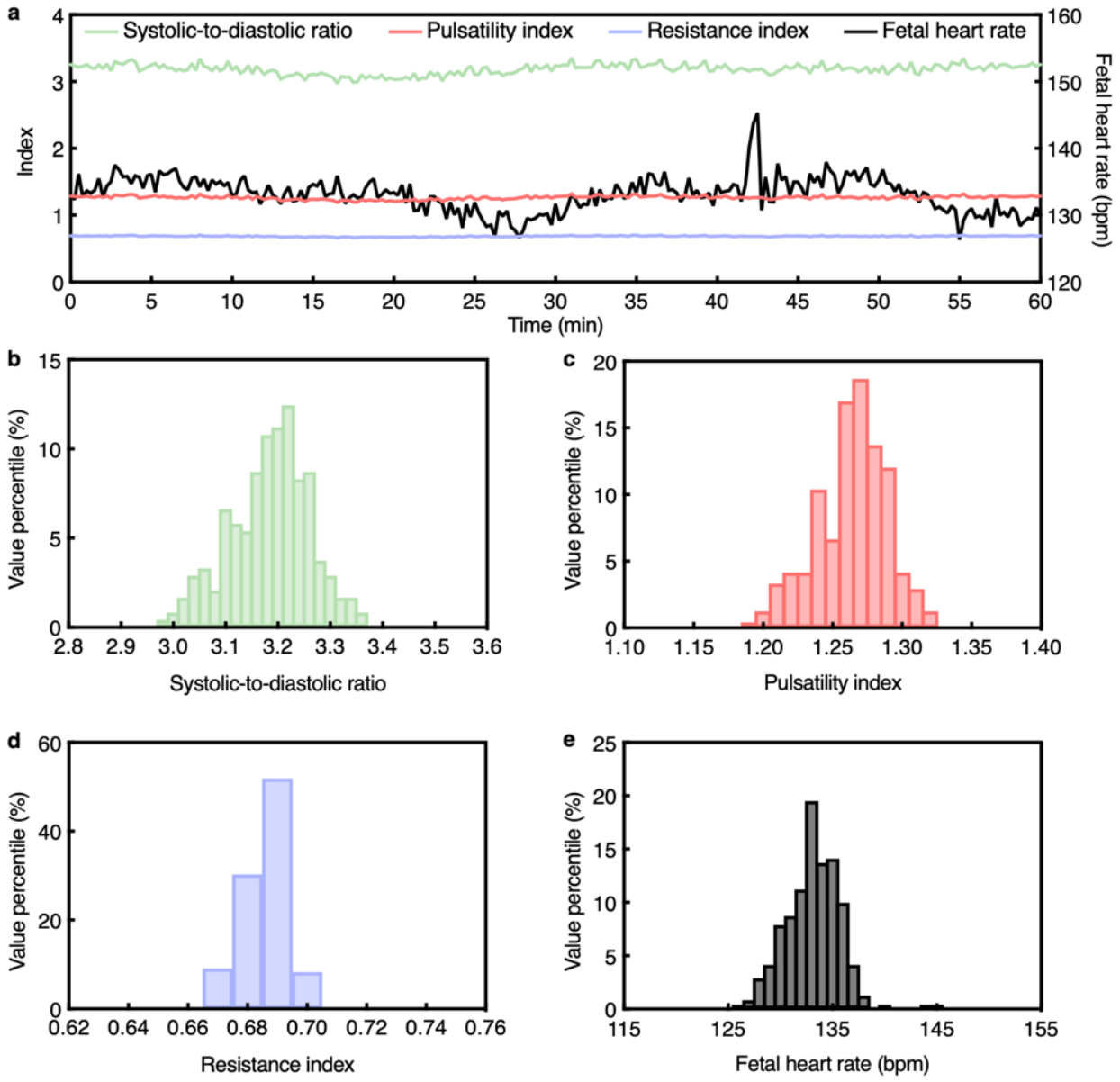
Supplementary Fig. 40 | Continuous fetal monitoring using the UPatch in Participant #5. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



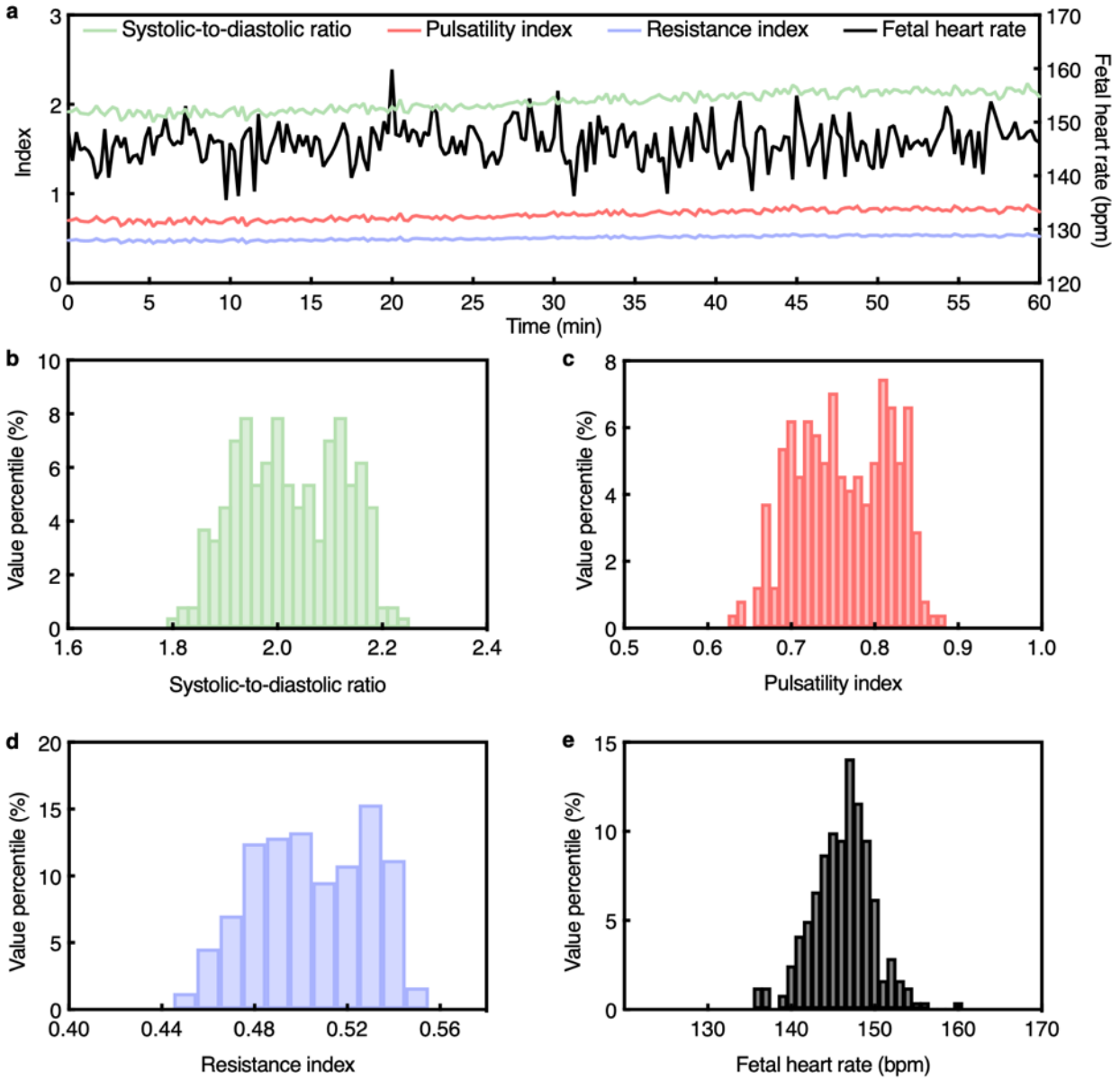
Supplementary Fig. 41 | Continuous fetal monitoring using the UPatch in Participant #6. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



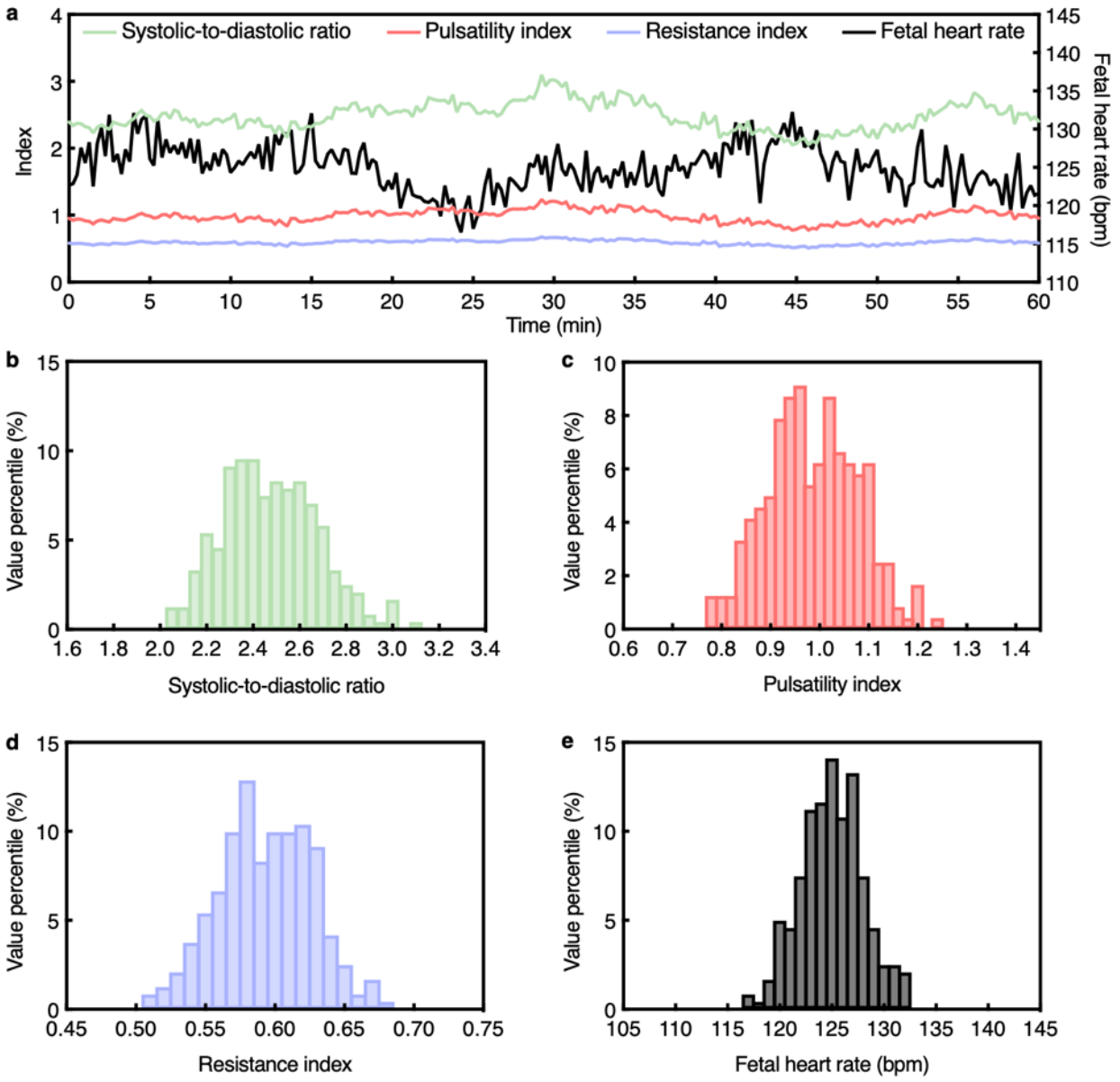
Supplementary Fig. 42 | Continuous fetal monitoring using the UPatch in Participant #7. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



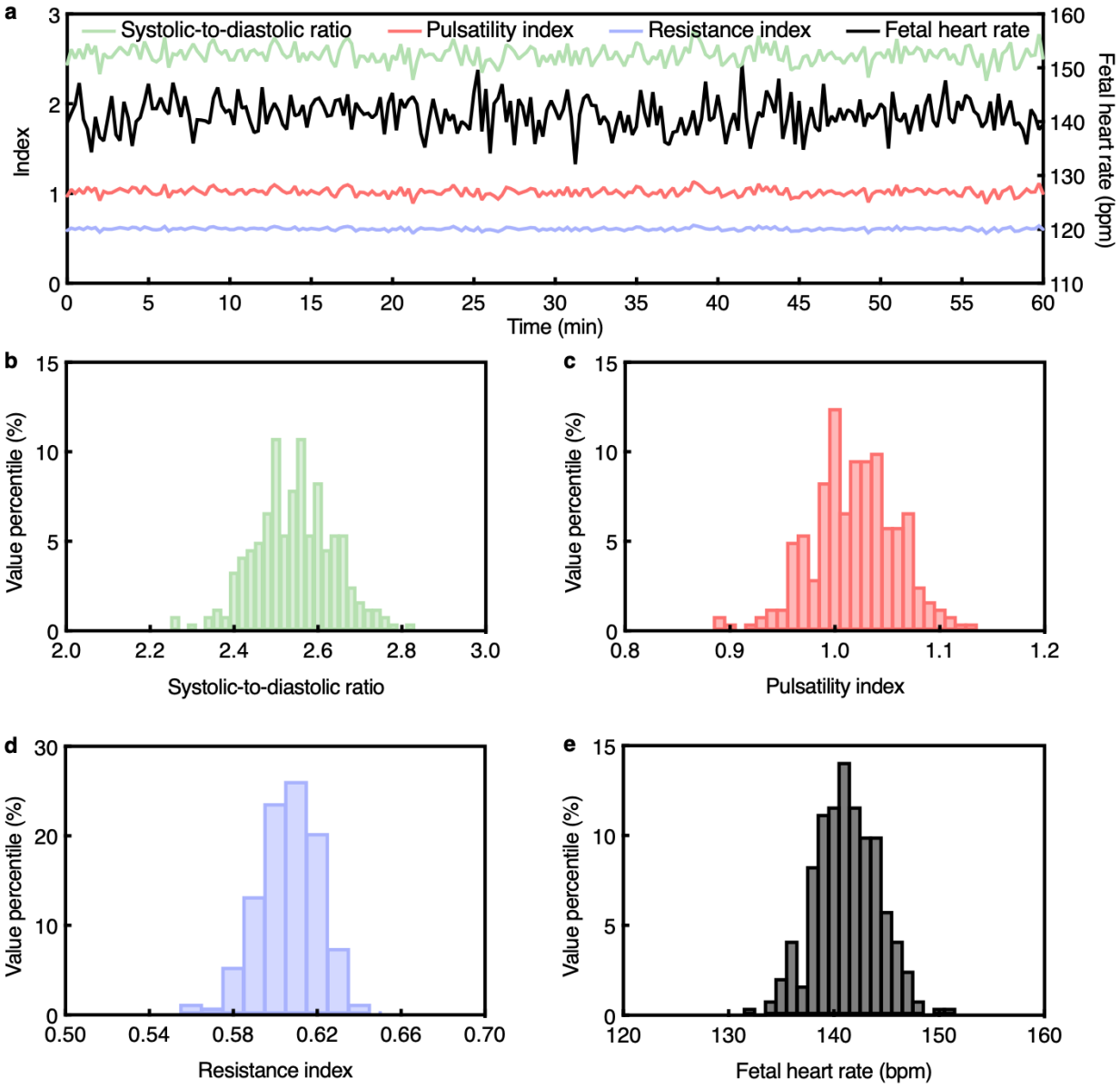
Supplementary Fig. 43 | Continuous fetal monitoring using the UPatch in Participant #8. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



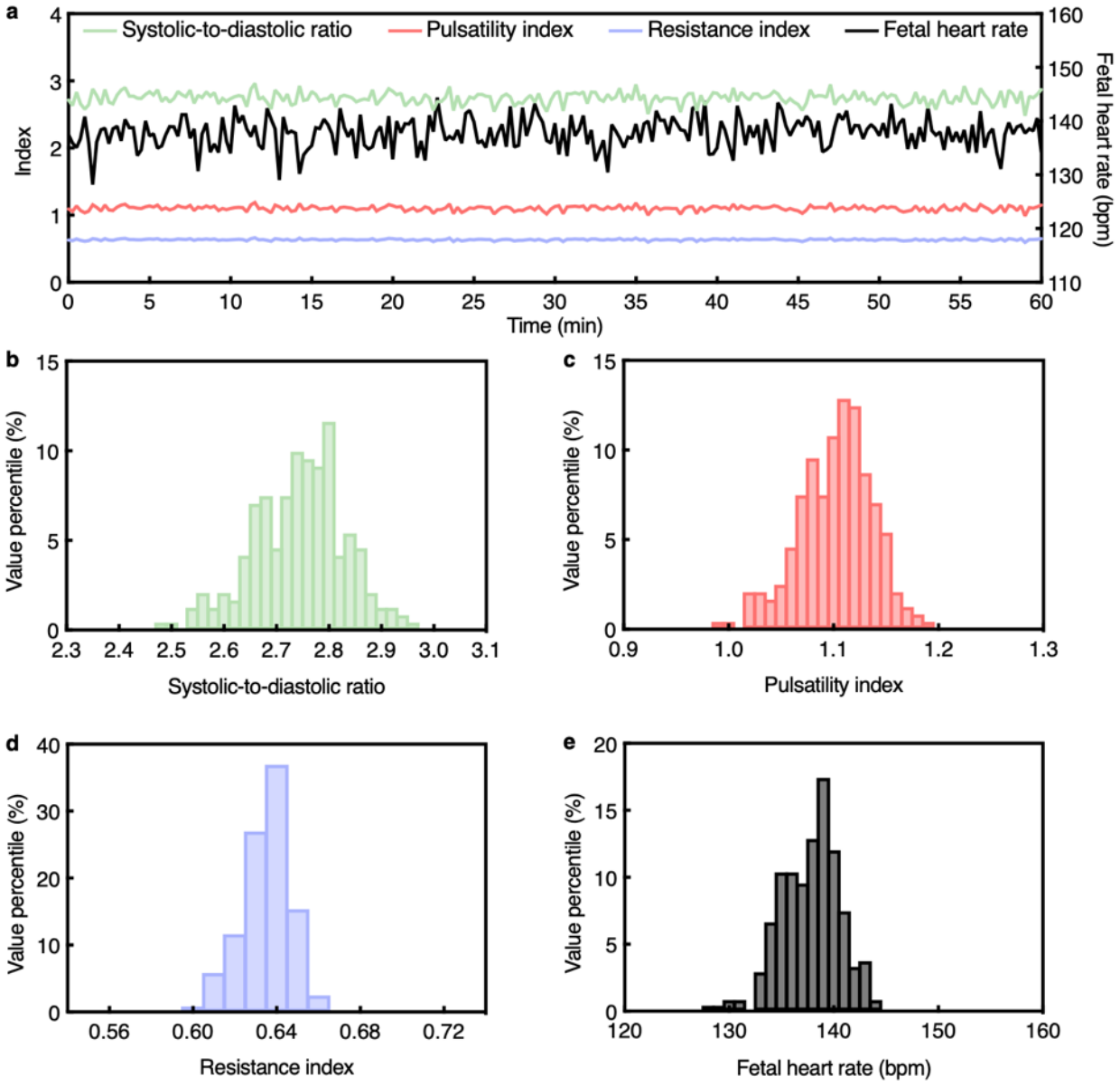
Supplementary Fig. 44 | Continuous fetal monitoring using the UPatch in Participant #9. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



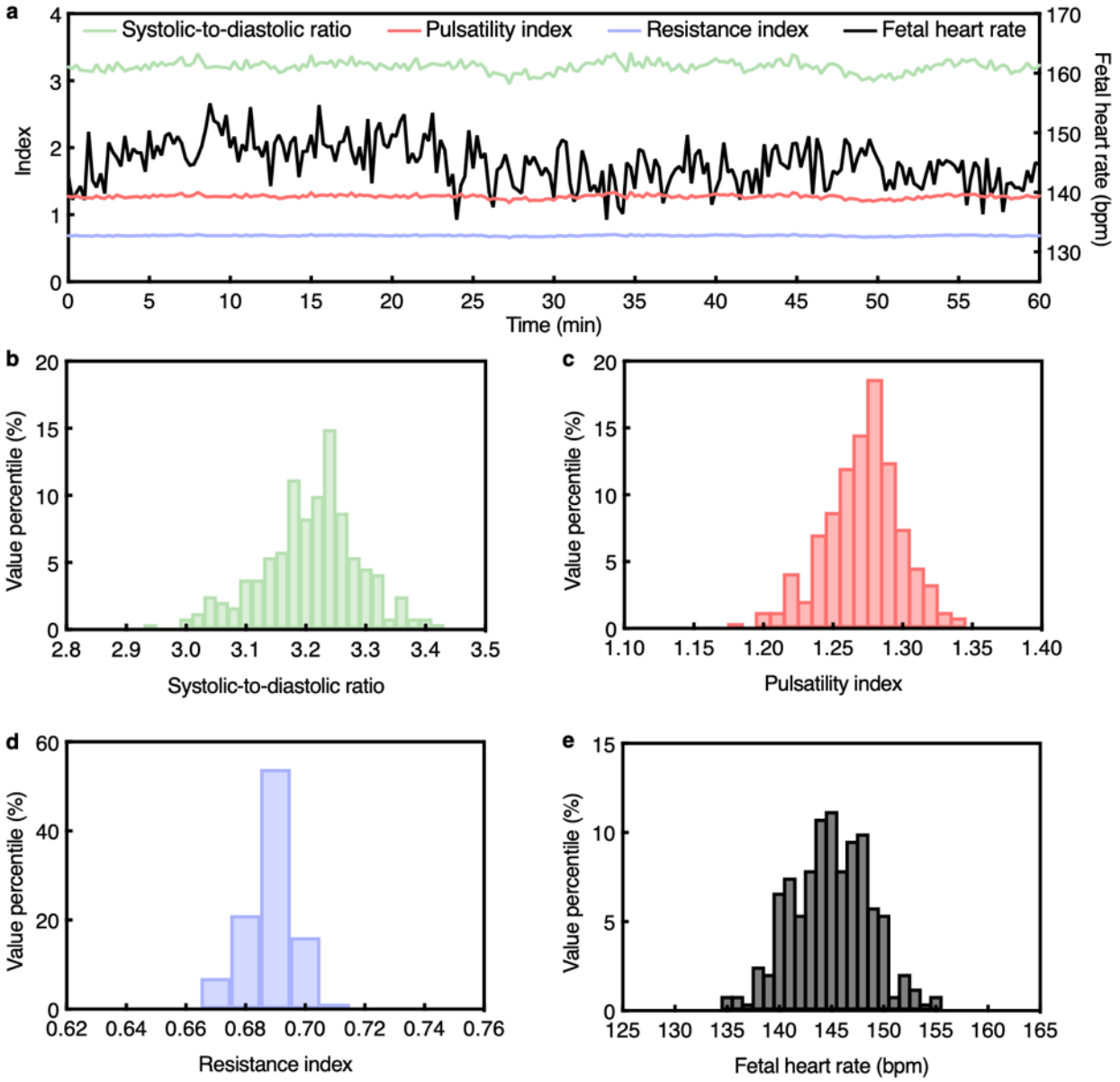
Supplementary Fig. 45 | Continuous fetal monitoring using the UPatch in Participant #10. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



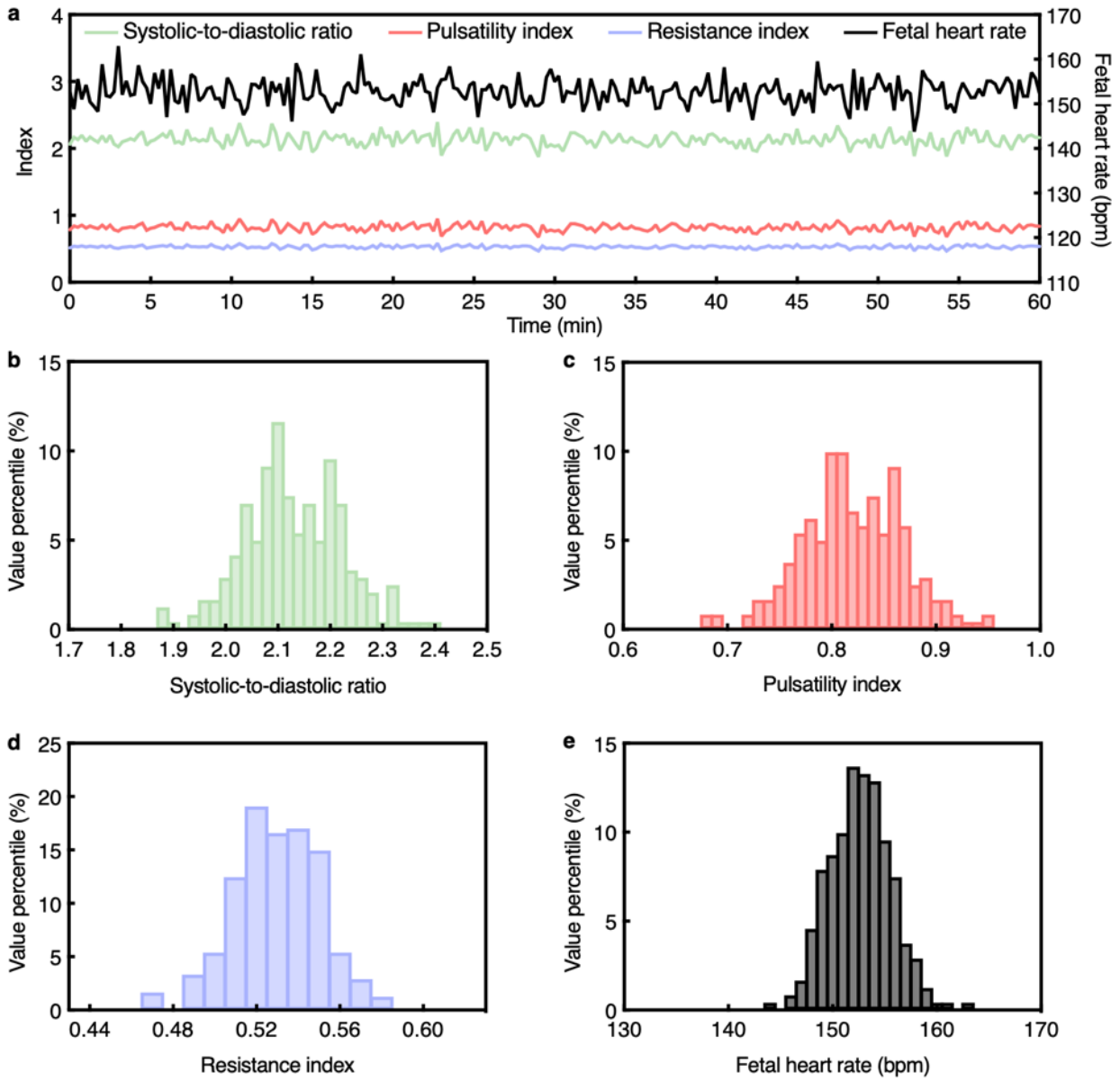
Supplementary Fig. 46 | Continuous fetal monitoring using the UPatch in Participant #11. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



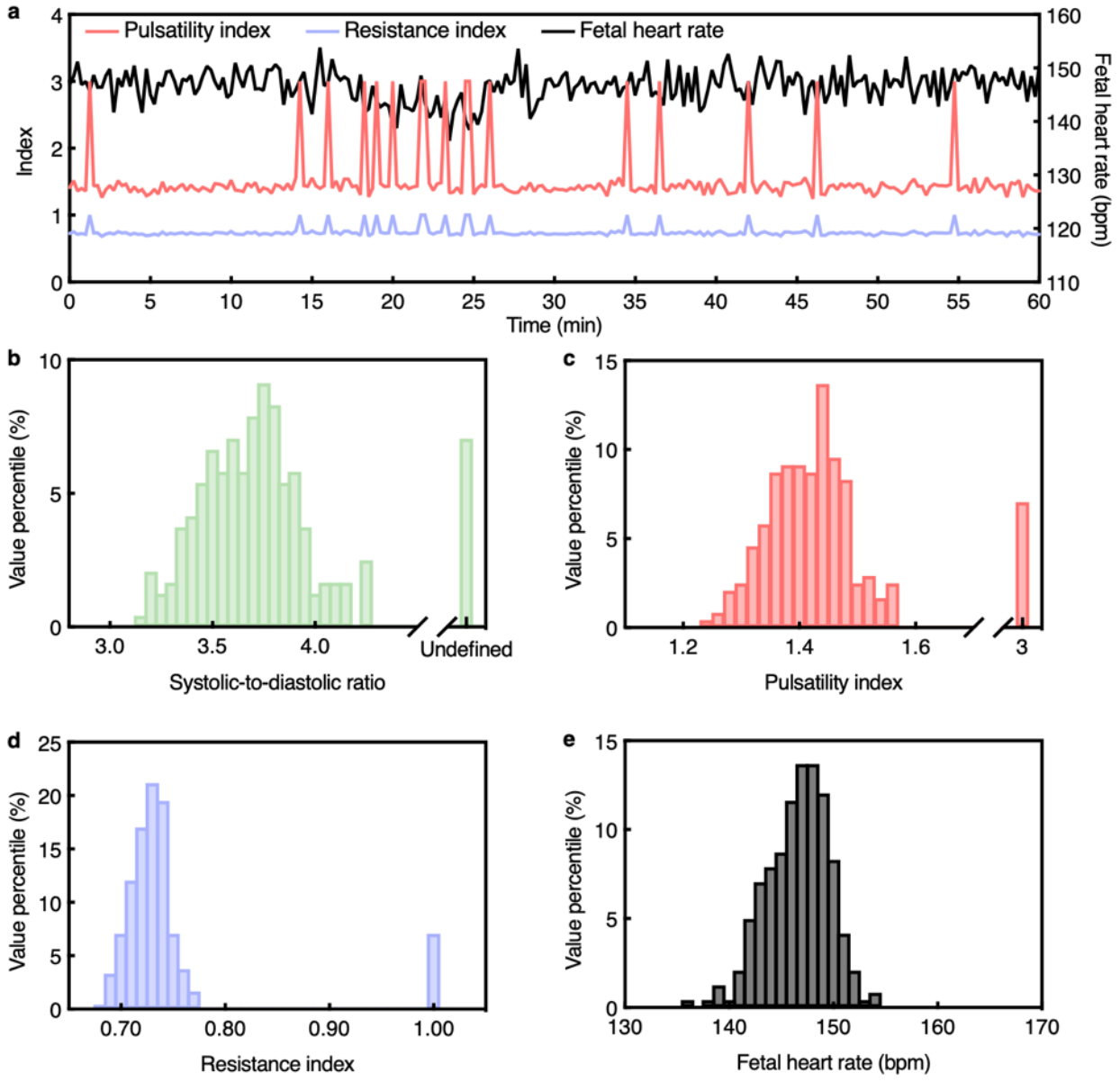
Supplementary Fig. 47 | Continuous fetal monitoring using the UPatch in Participant #12. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



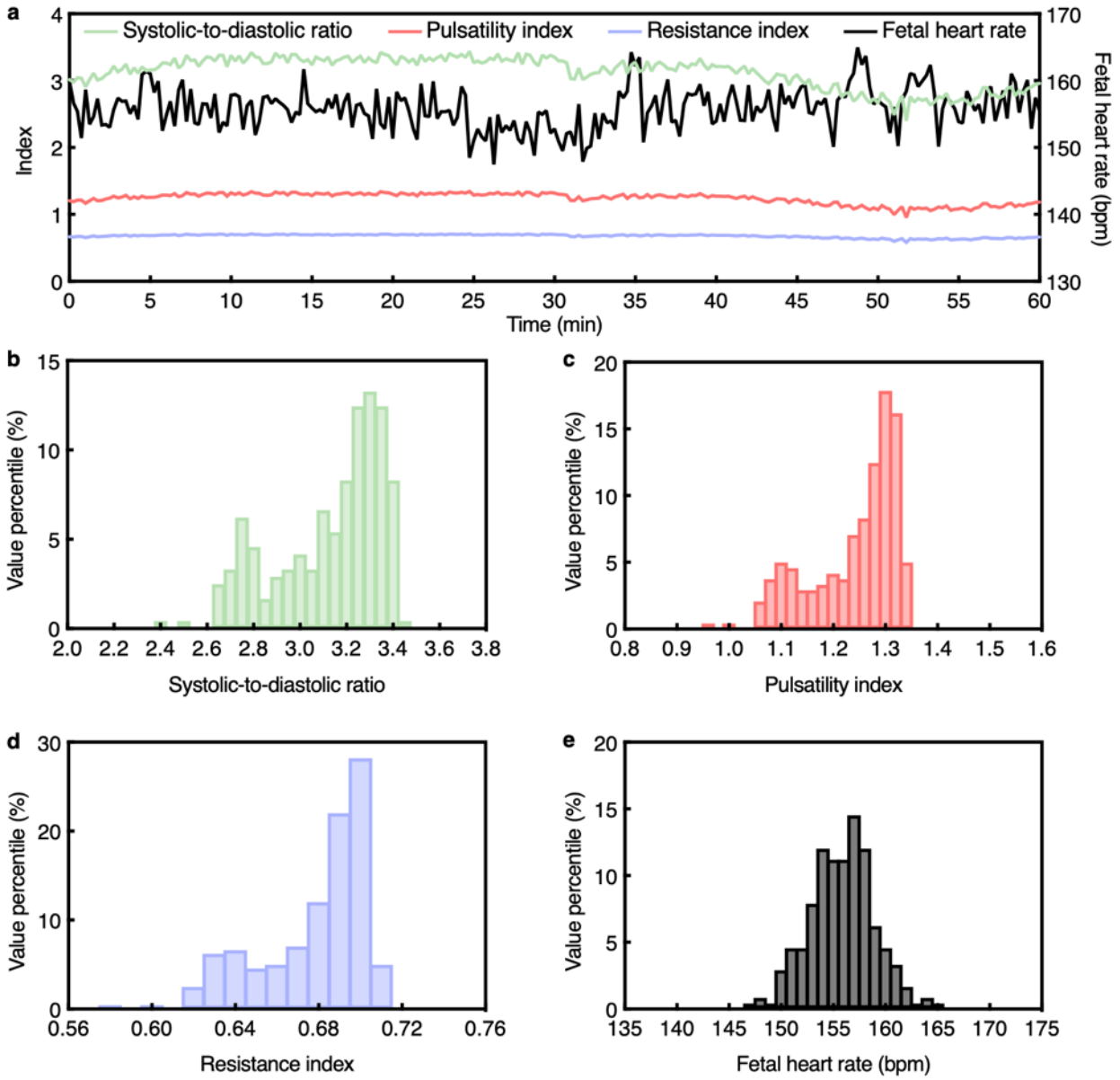
Supplementary Fig. 48 | Continuous fetal monitoring using the UPatch in Participant #13. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



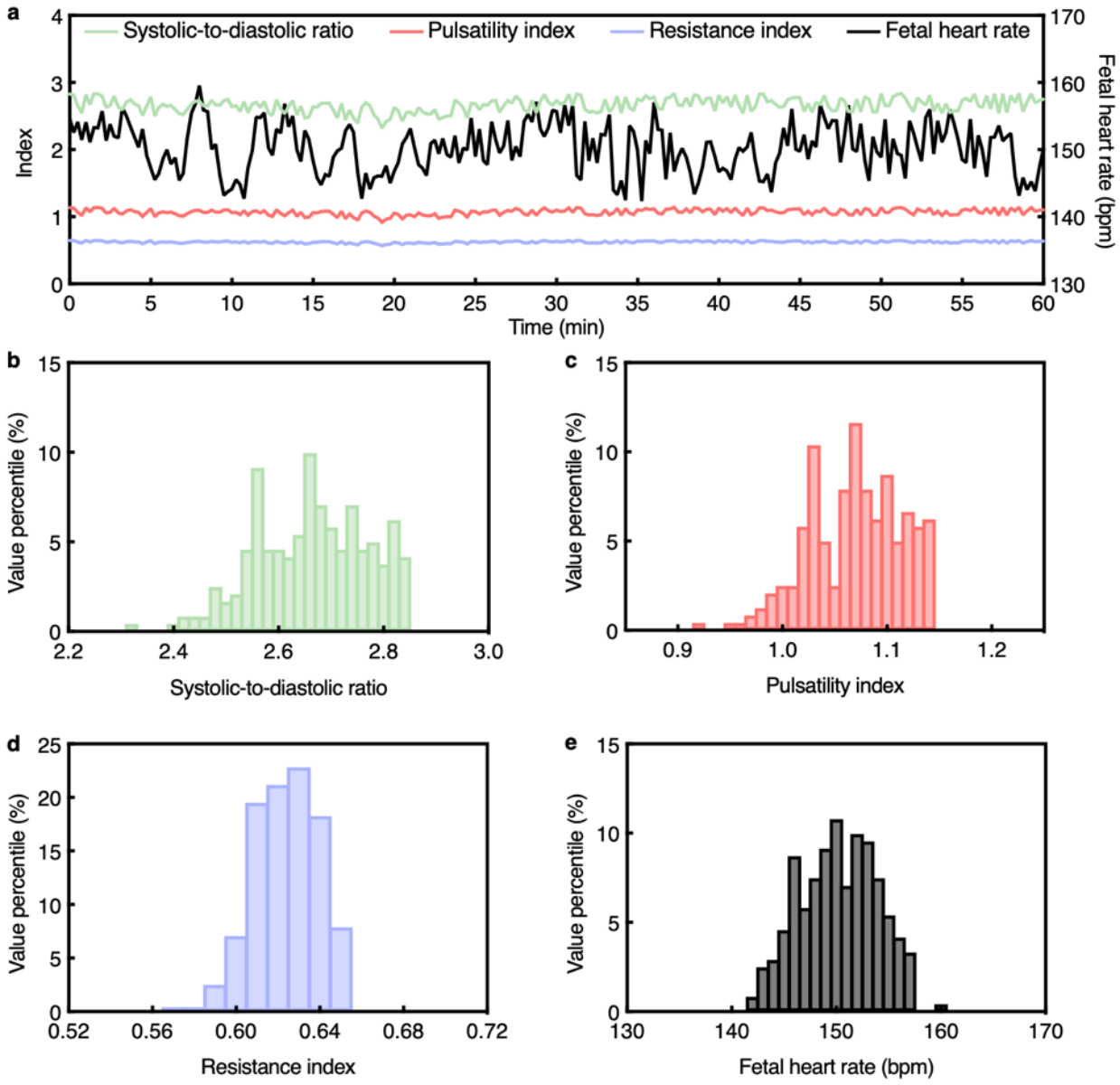
Supplementary Fig. 49 | Continuous fetal monitoring using the UPatch in Participant #14. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



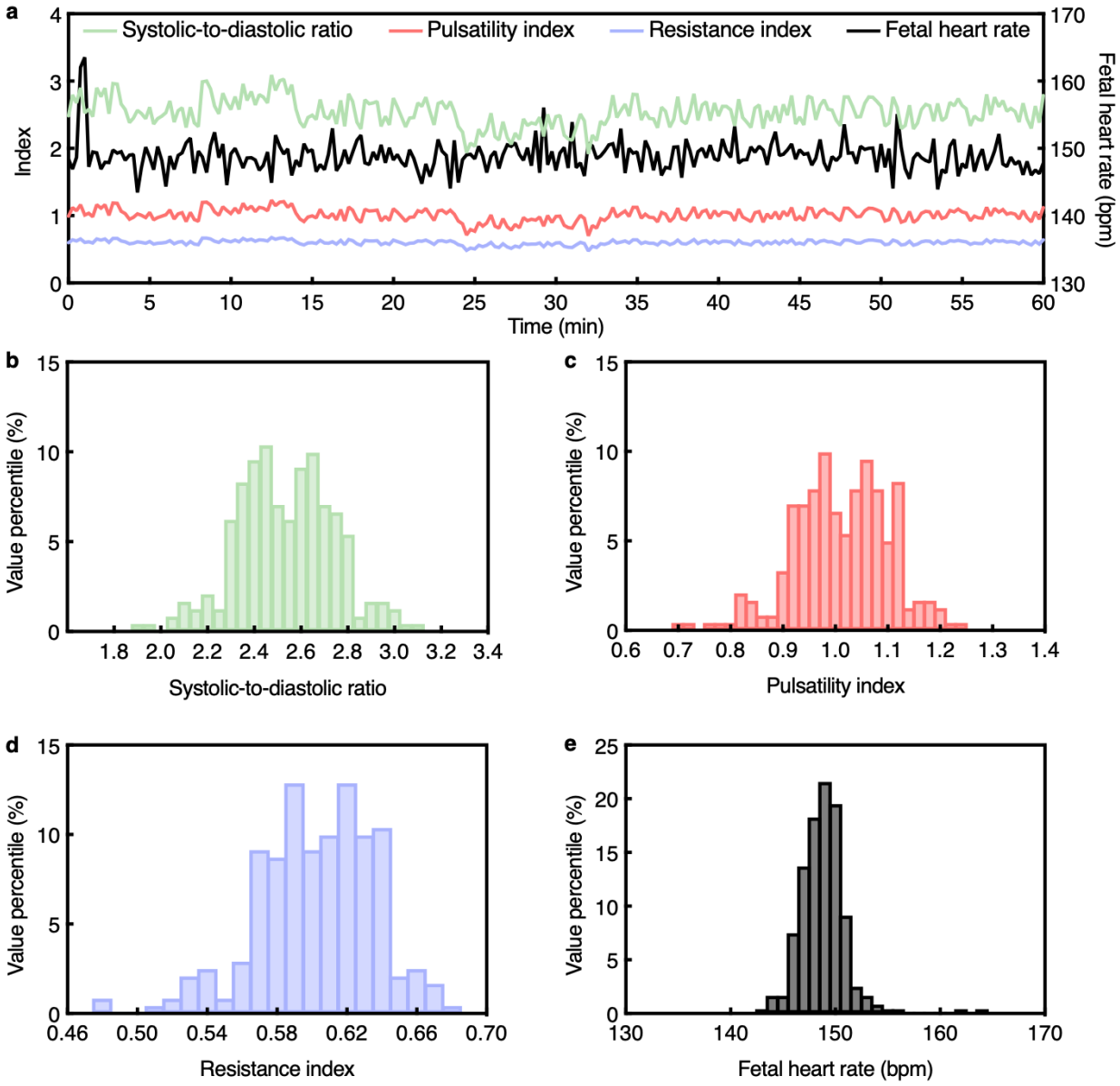
Supplementary Fig. 50 | Continuous fetal monitoring using the UPatch in Participant #15. a, Time series data from the 1-h continuous monitoring session. Pulsatility index and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. The systolic-to-diastolic ratio was not plotted in the time series due to the presence of undefined values. The large peaks correspond to periods of absent end diastolic flow, during which the systolic-to-diastolic ratio is undefined, the pulsatility index is 3.0, and the resistance index is 1.0. Such absence of end diastolic flow is clinically relevant because it is associated with abnormal feto-placental hemodynamics and can indicate evolving intrauterine growth restriction, whereas healthy fetuses typically maintain forward end diastolic flow[49,50]. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



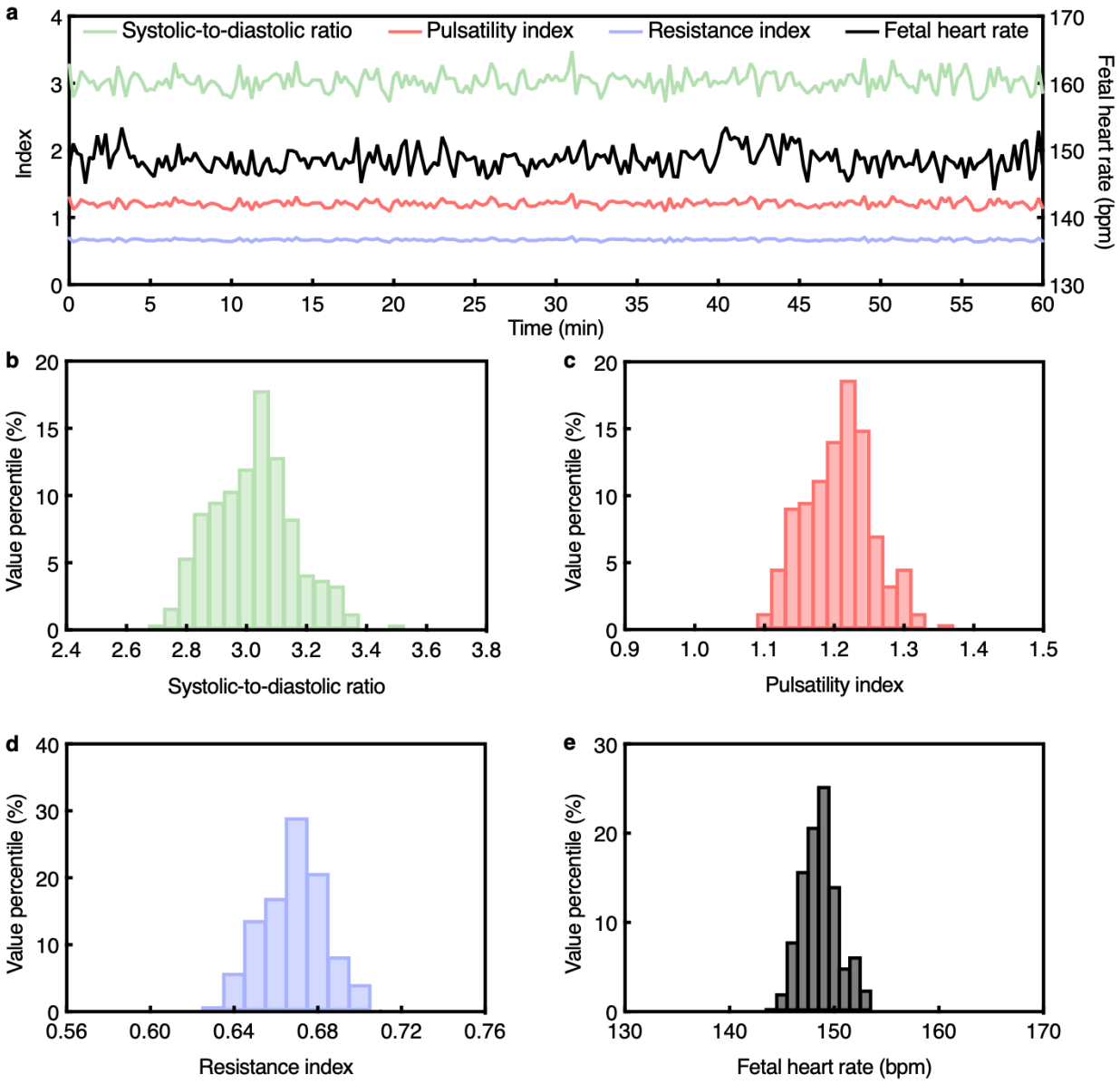
Supplementary Fig. 51 | Continuous fetal monitoring using the UPatch in Participant #16. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



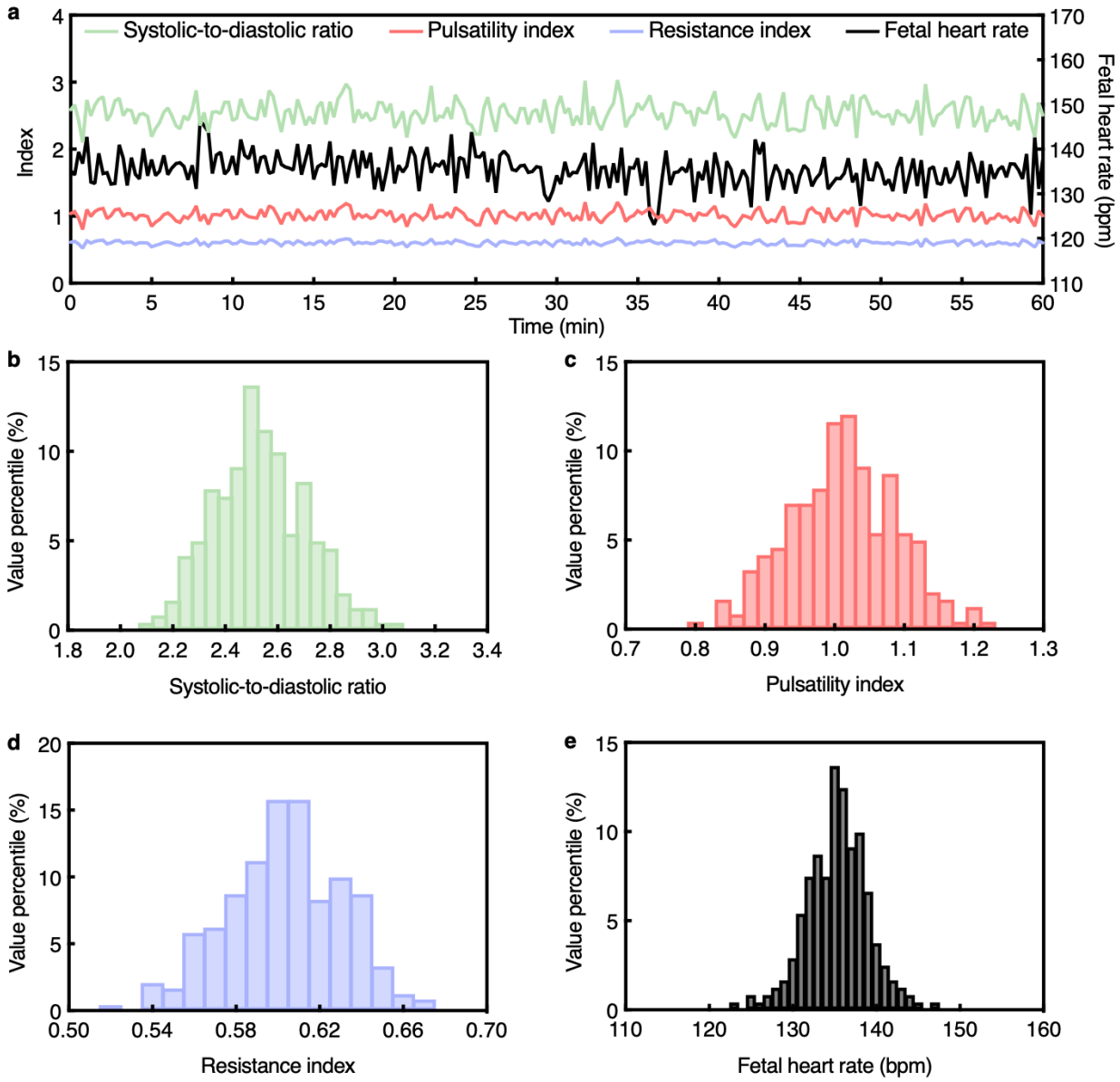
Supplementary Fig. 52 | Continuous fetal monitoring using the UPatch in Participant #17. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



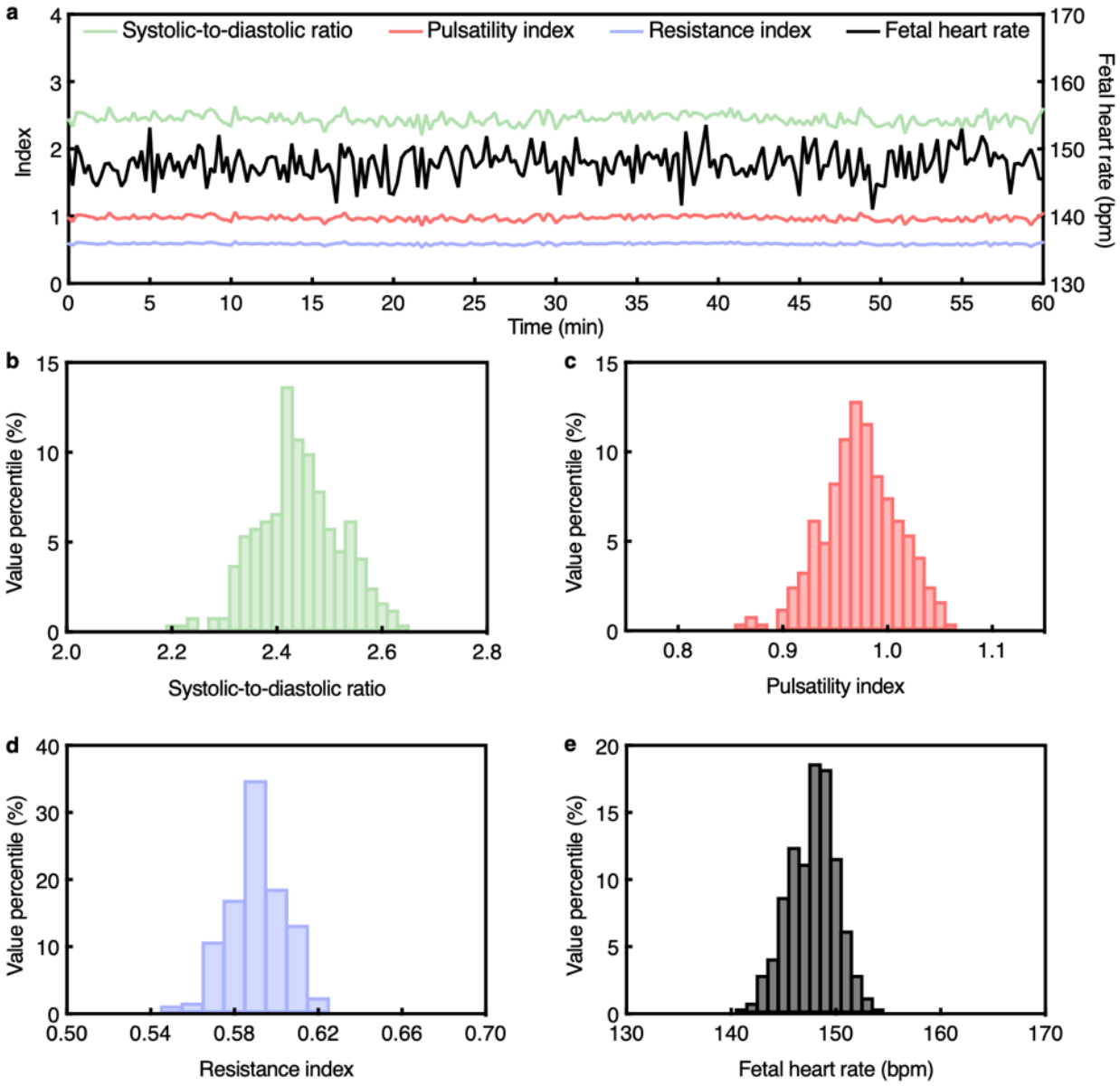
Supplementary Fig. 53 | Continuous fetal monitoring using the UPatch in Participant #18. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



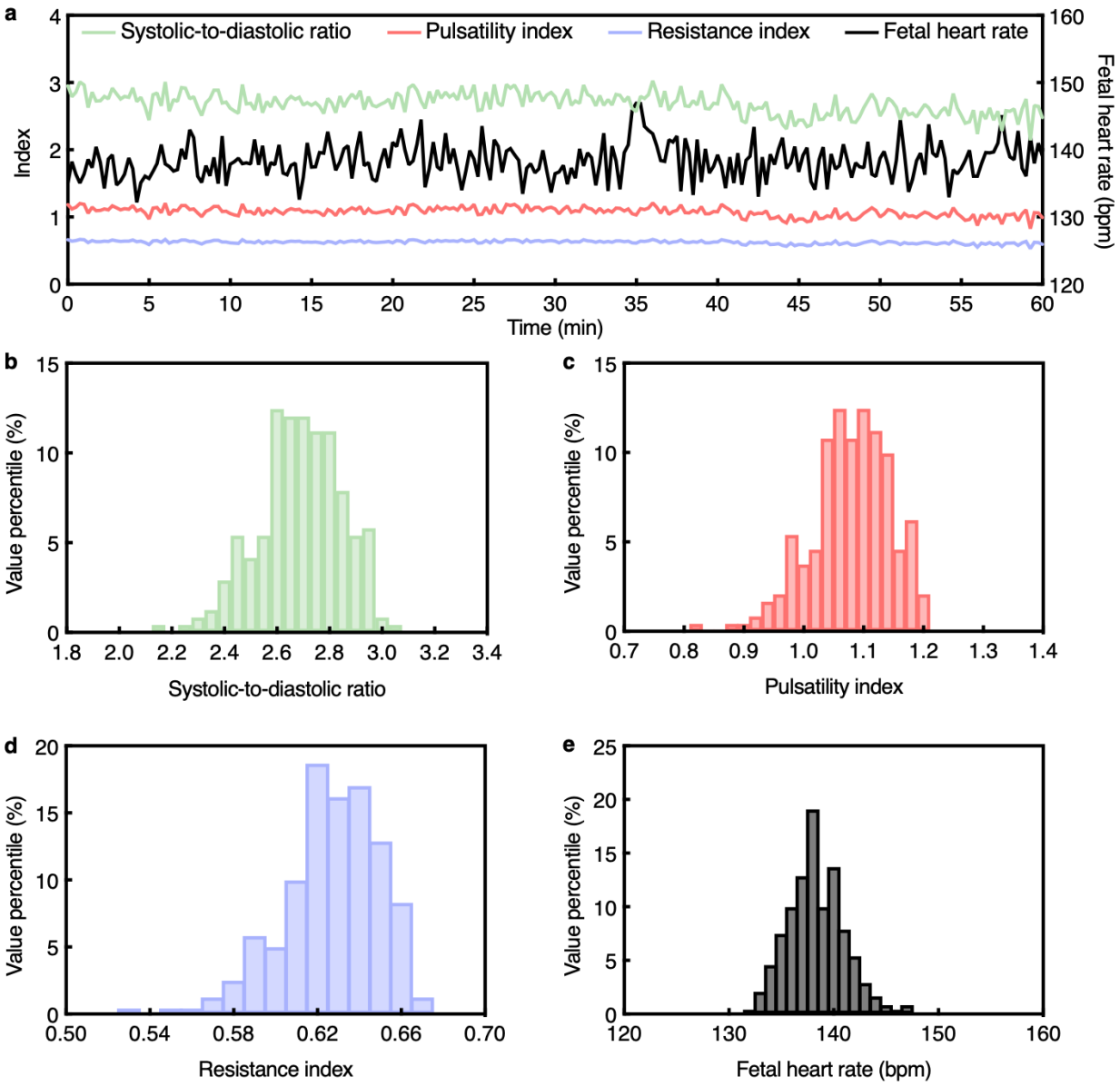
Supplementary Fig. 54 | Continuous fetal monitoring using the UPatch in Participant #19. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



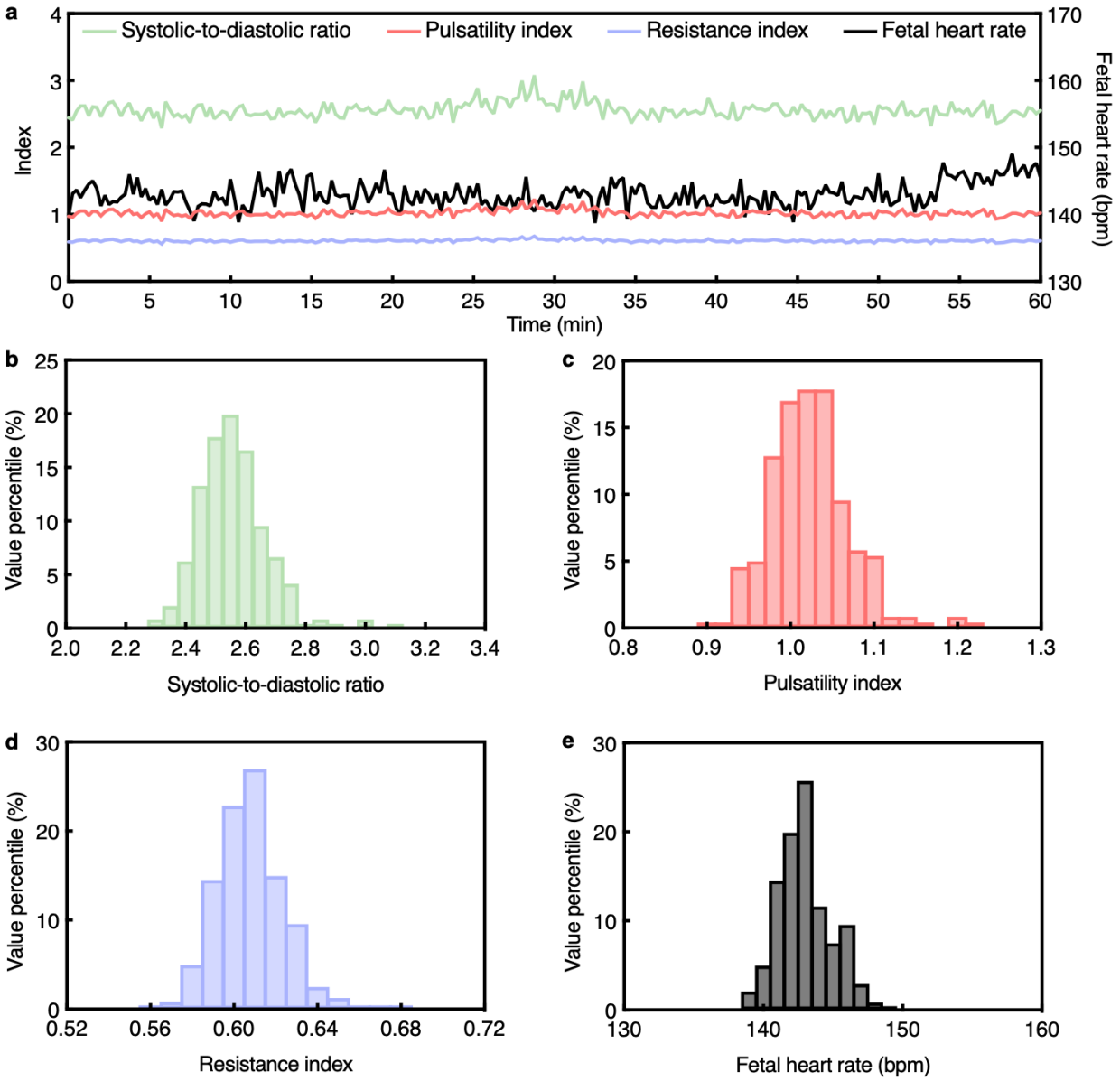
Supplementary Fig. 55 | Continuous fetal monitoring using the UPatch in Participant #20. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



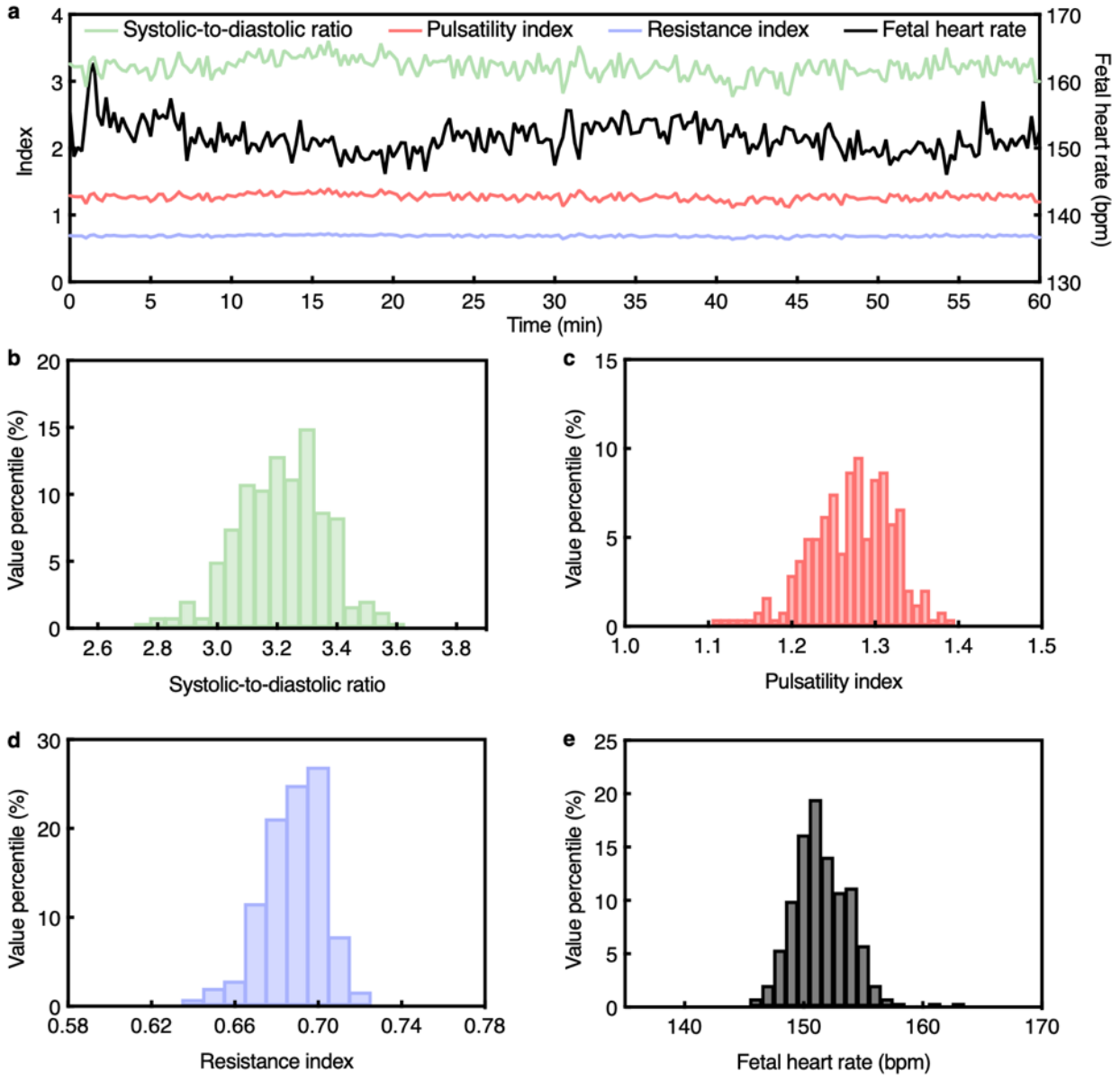
Supplementary Fig. 56 | Continuous fetal monitoring using the UPatch in Participant #21. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



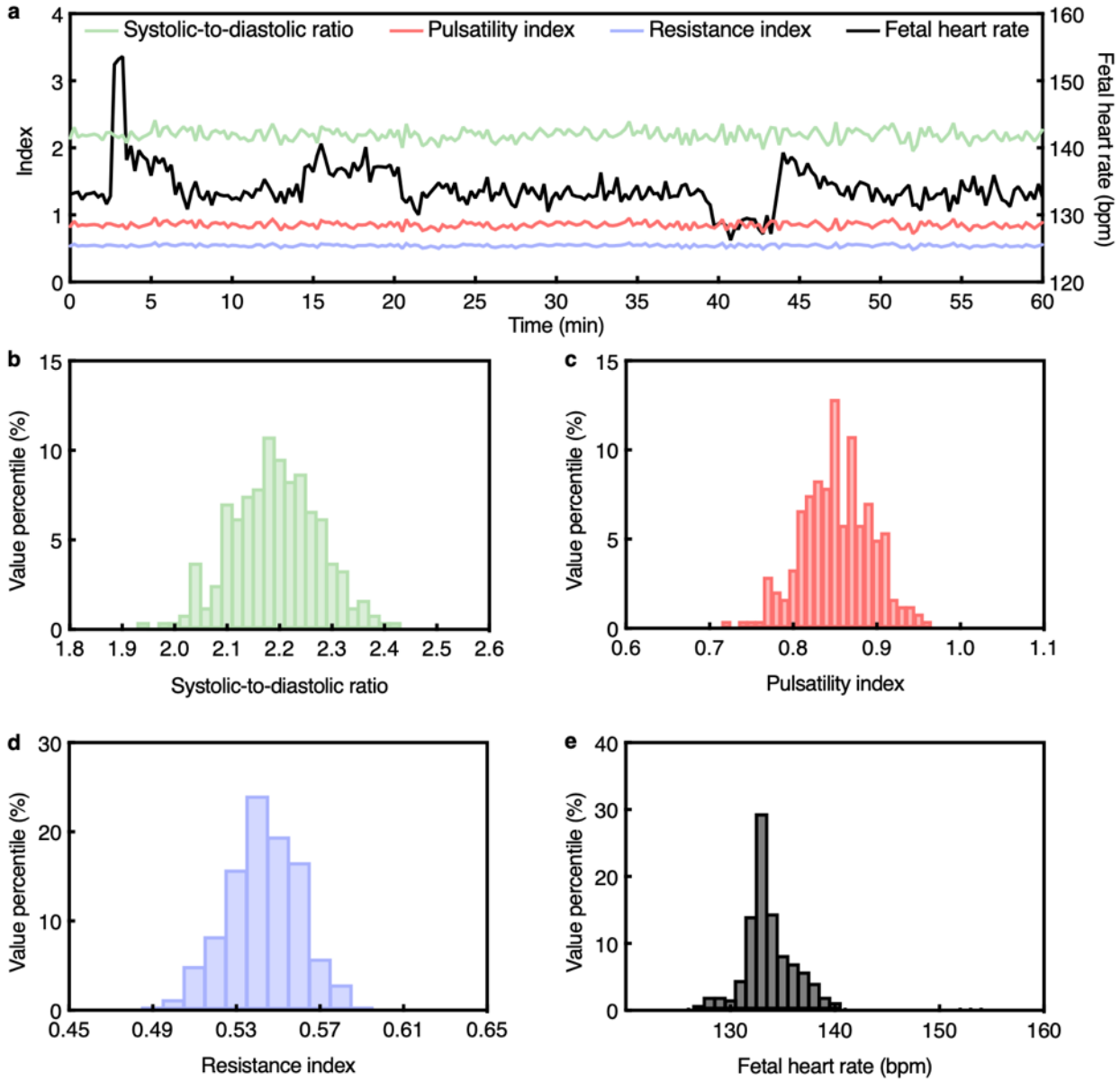
Supplementary Fig. 57 | Continuous fetal monitoring using the UPatch in Participant #22. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



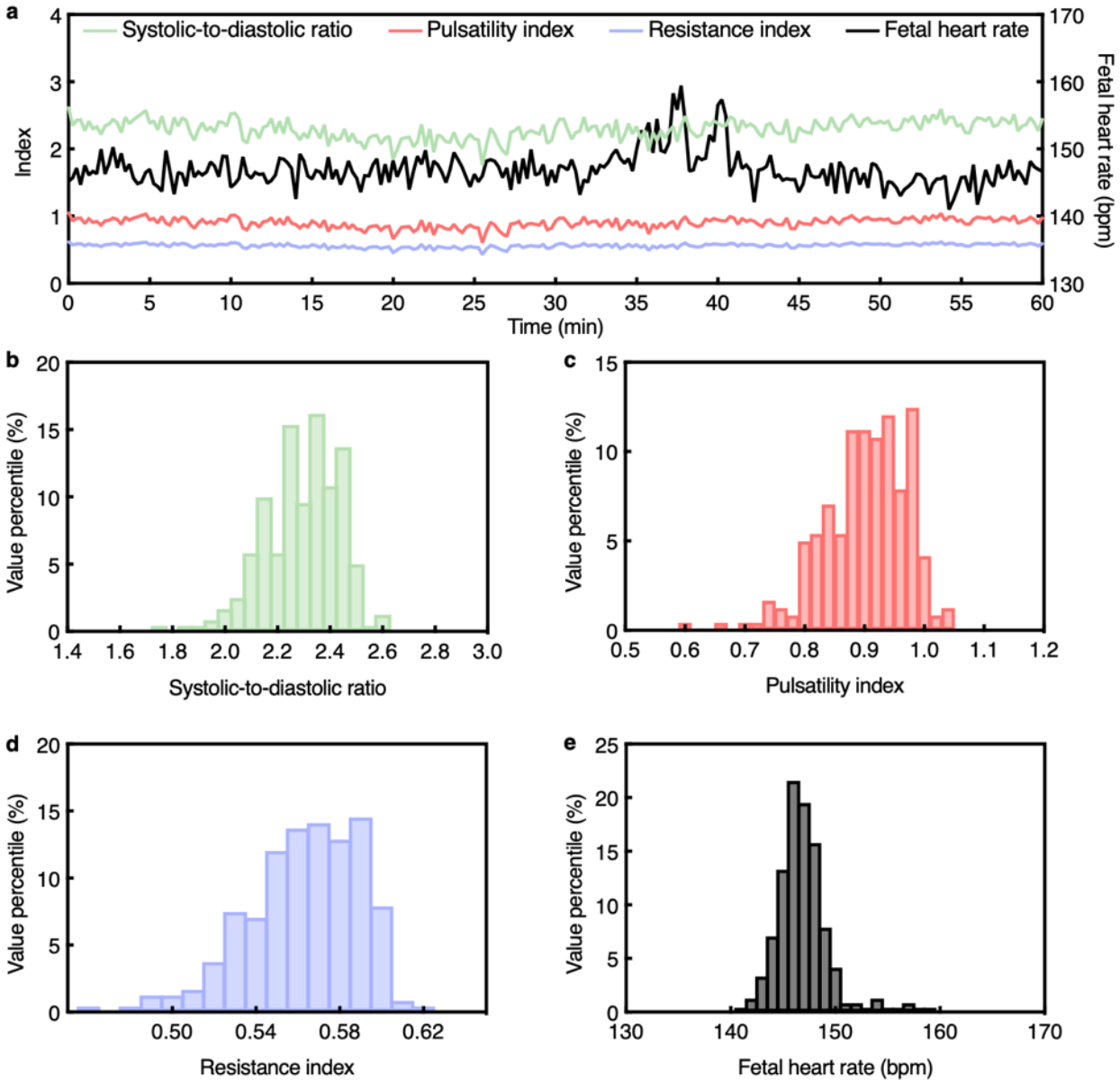
Supplementary Fig. 58 | Continuous fetal monitoring using the UPatch in Participant #23. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



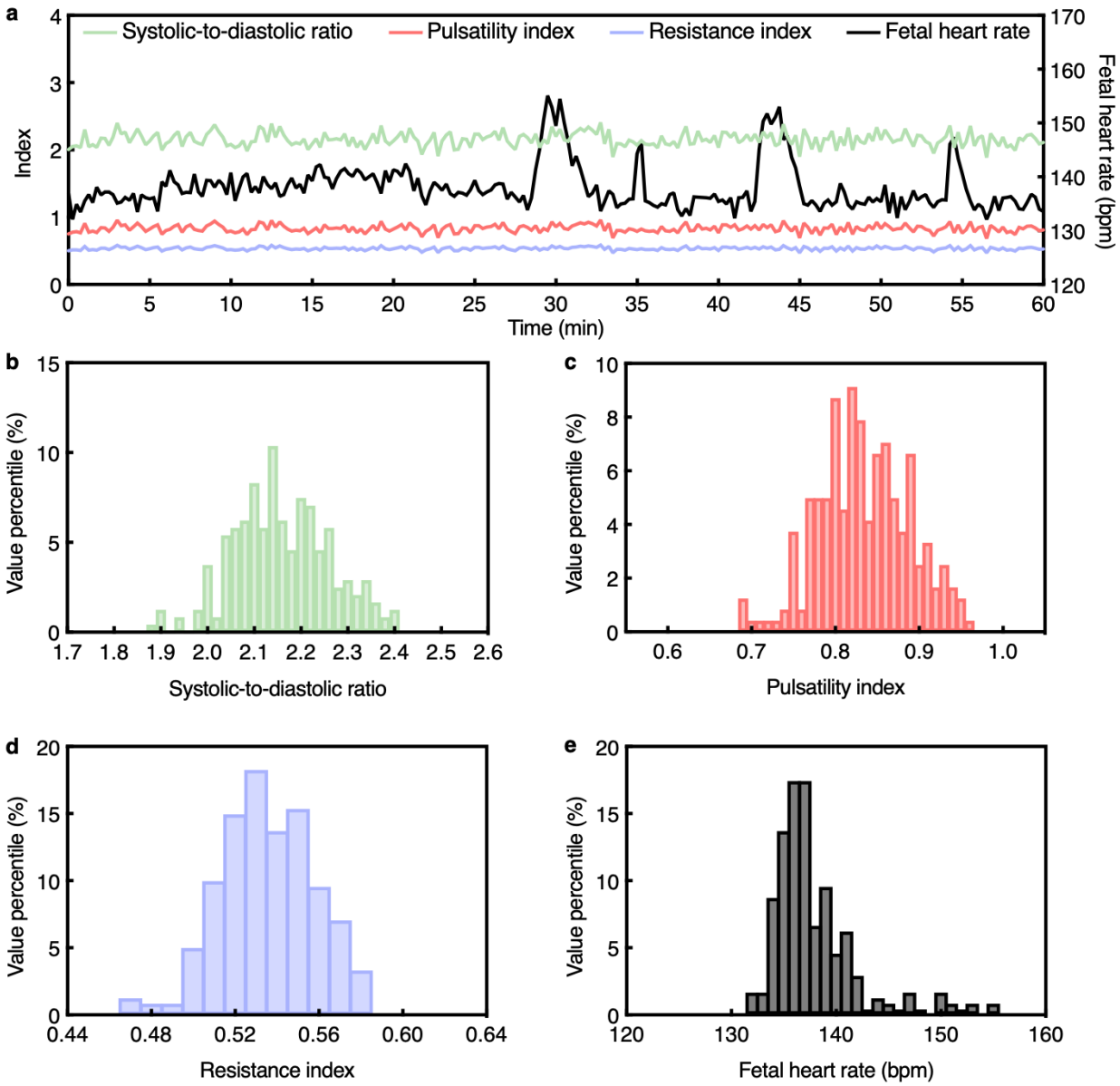
Supplementary Fig. 59 | Continuous fetal monitoring using the UPatch in Participant #24. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



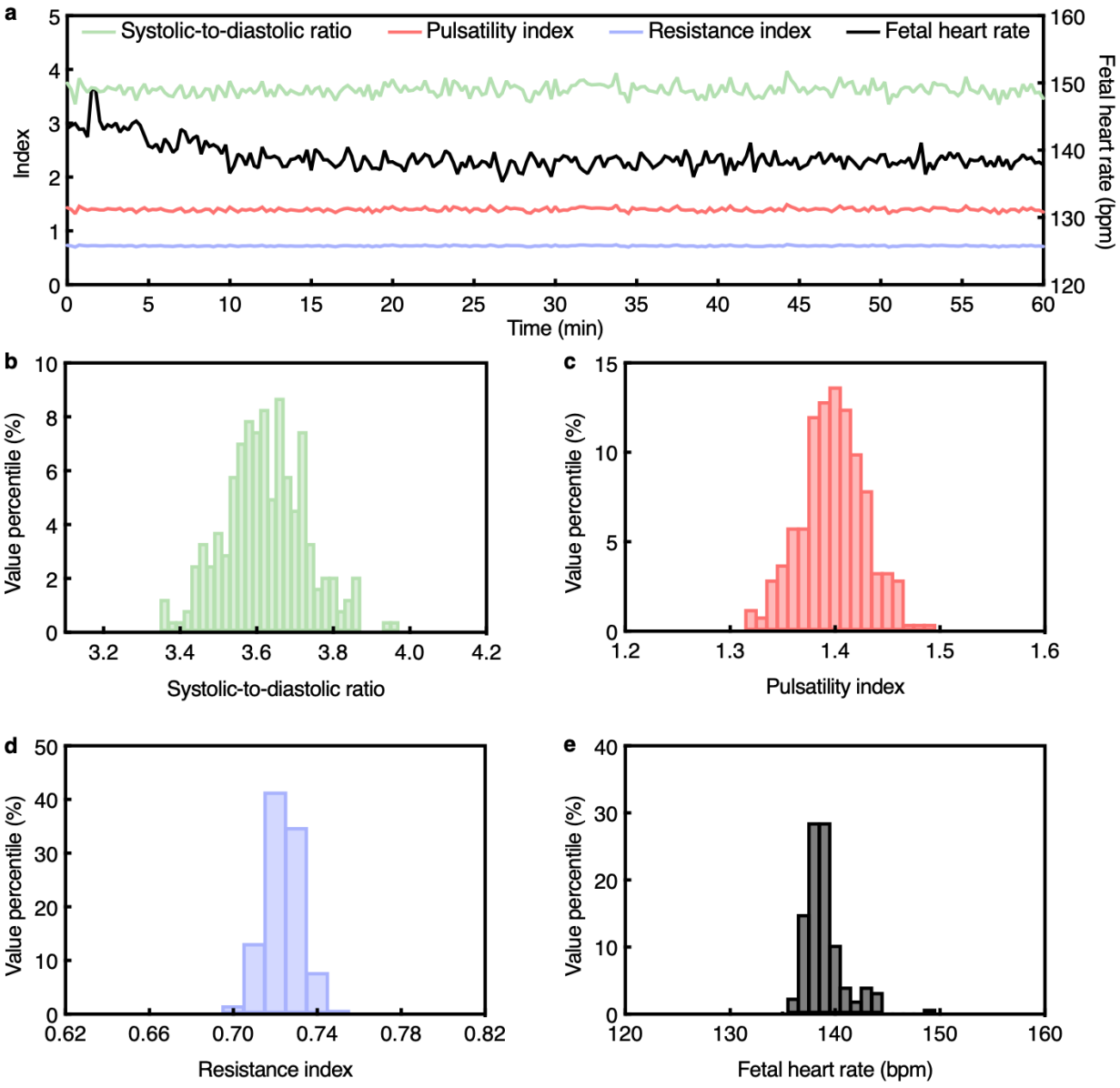
Supplementary Fig. 60 | Continuous fetal monitoring using the UPatch in Participant #25. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



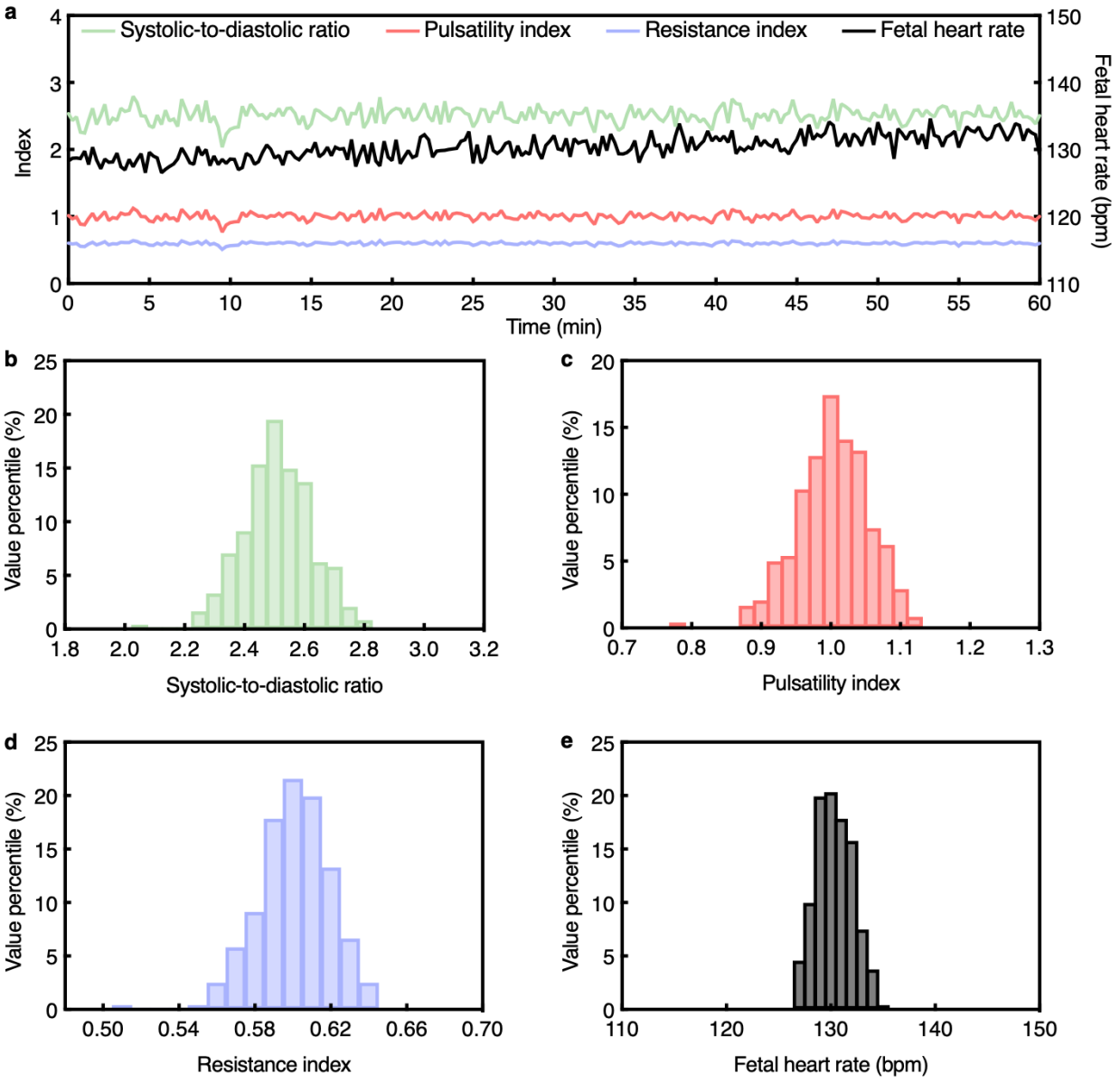
Supplementary Fig. 61 | Continuous fetal monitoring using the UPatch in Participant #26. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



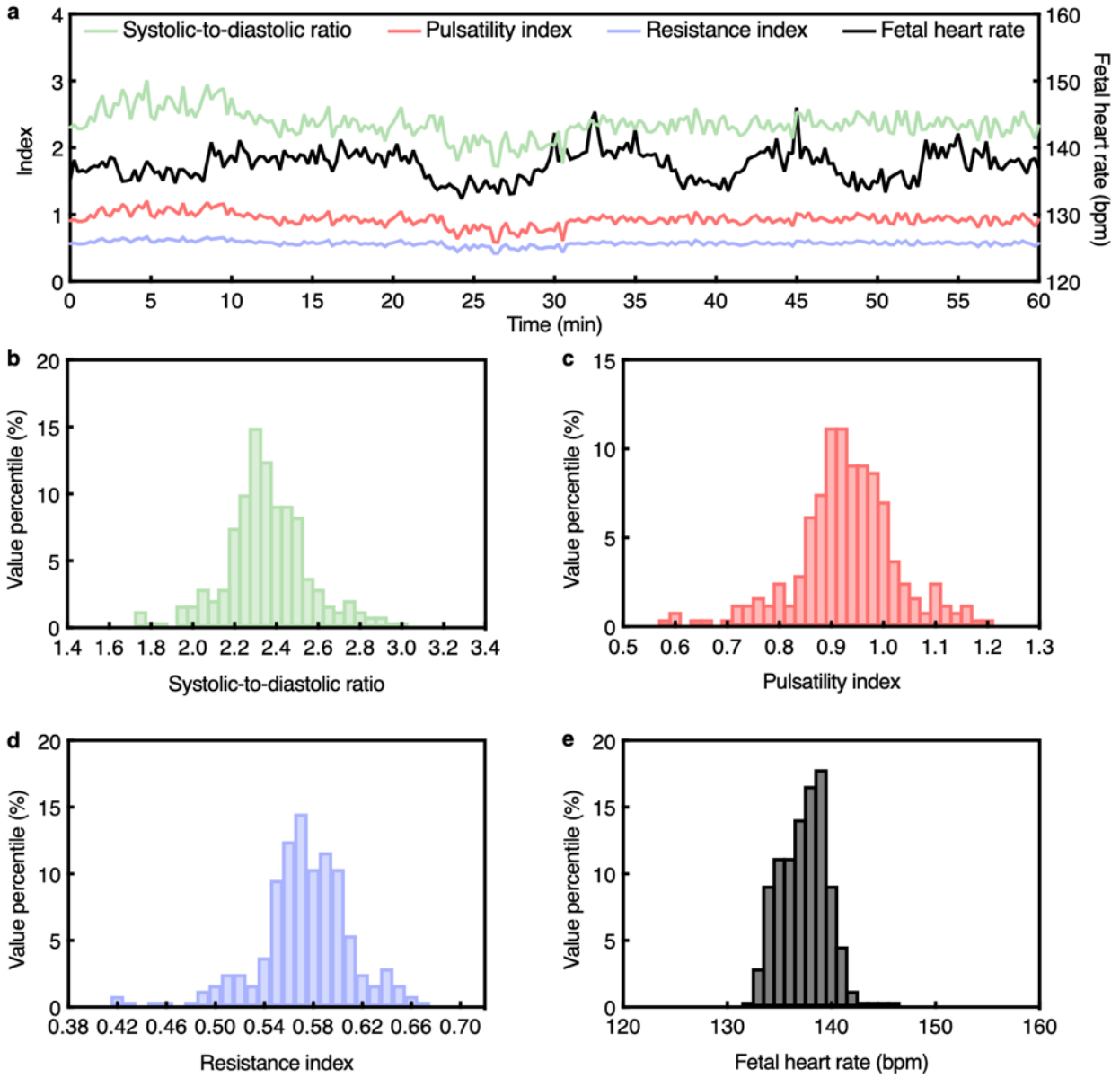
Supplementary Fig. 62 | Continuous fetal monitoring using the UPatch in Participant #27. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



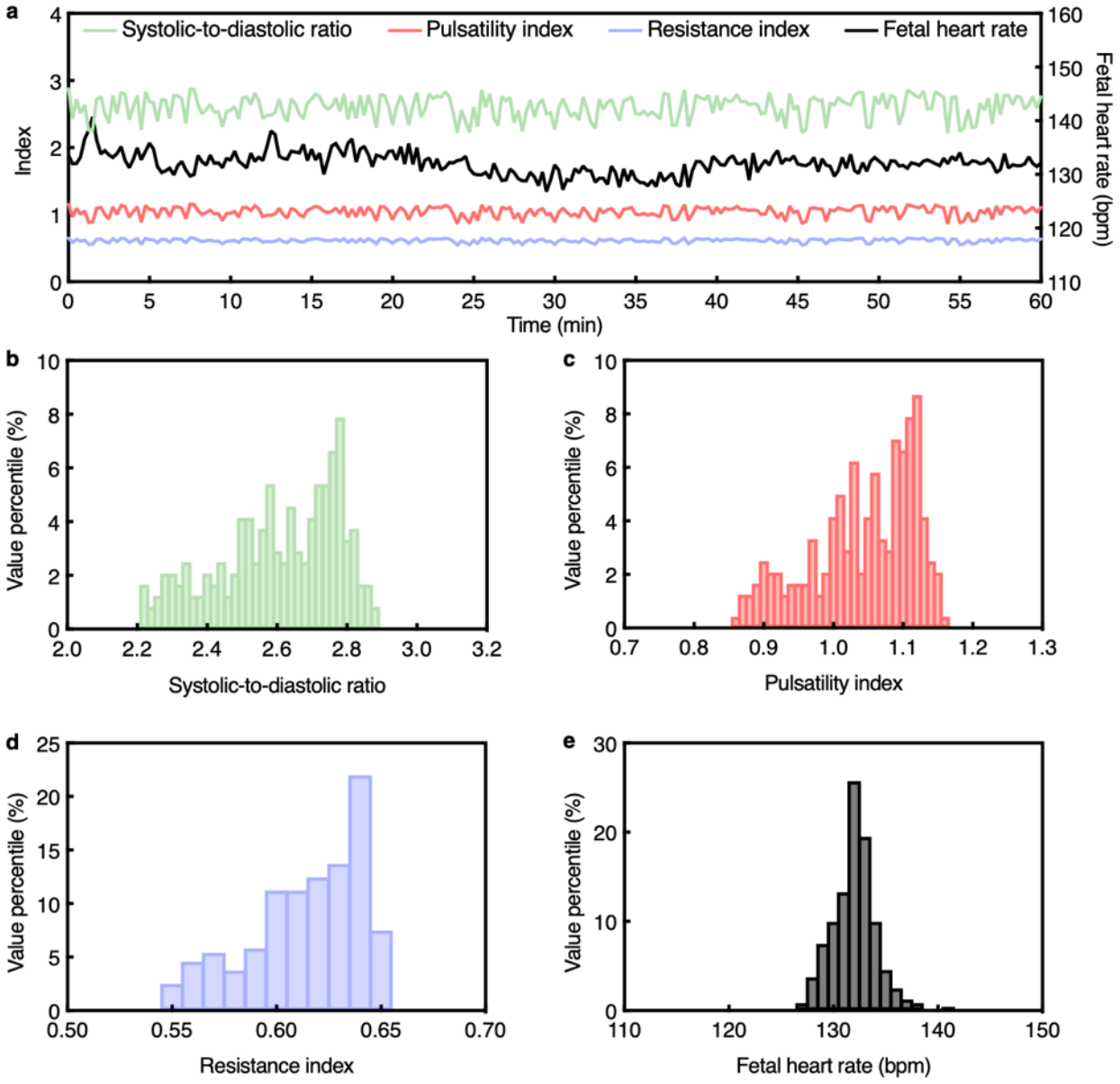
Supplementary Fig. 63 | Continuous fetal monitoring using the UPatch in Participant #28. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



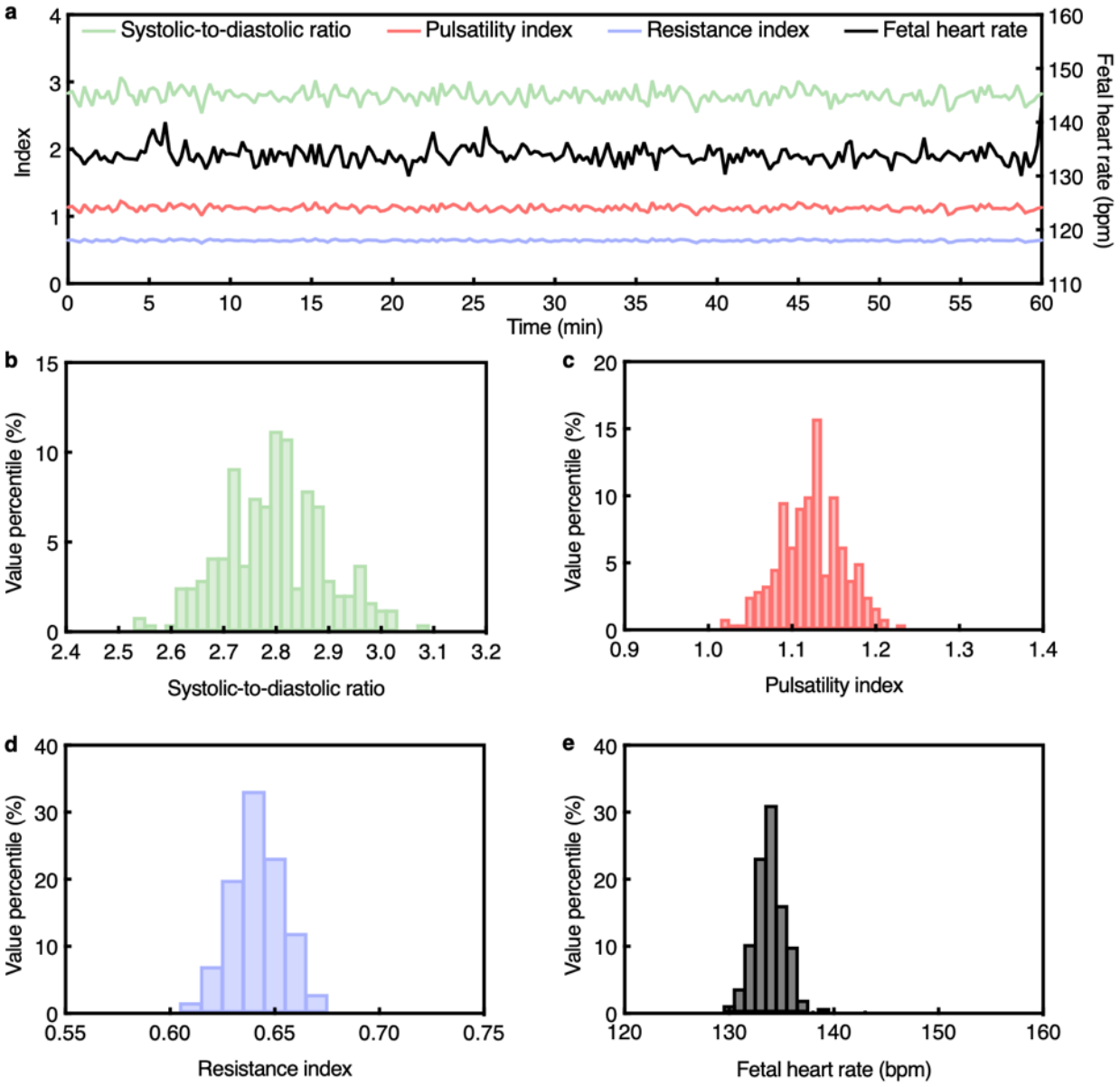
Supplementary Fig. 64 | Continuous fetal monitoring using the UPatch in Participant #29. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



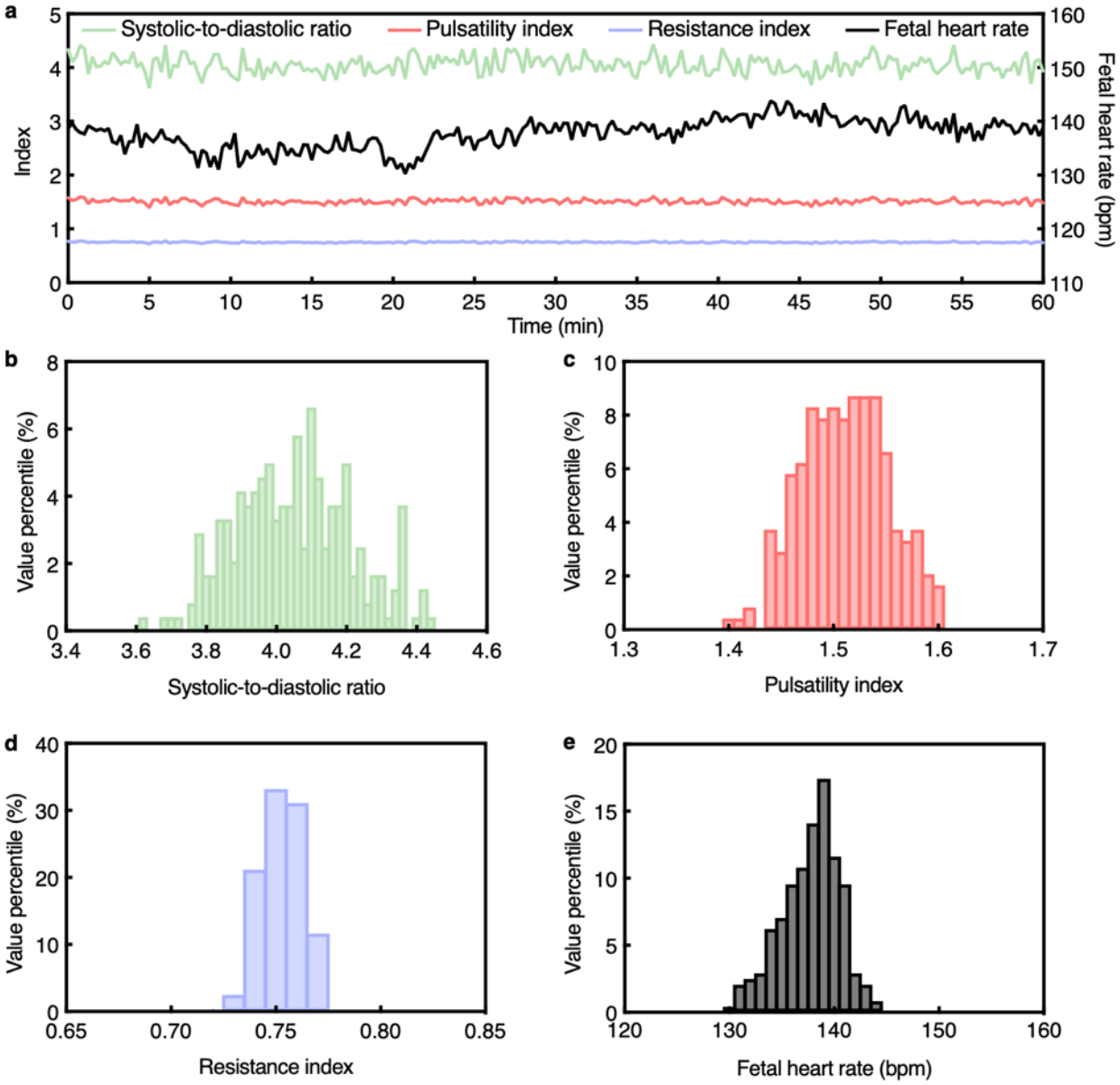
Supplementary Fig. 65 | Continuous fetal monitoring using the UPatch in Participant #30. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



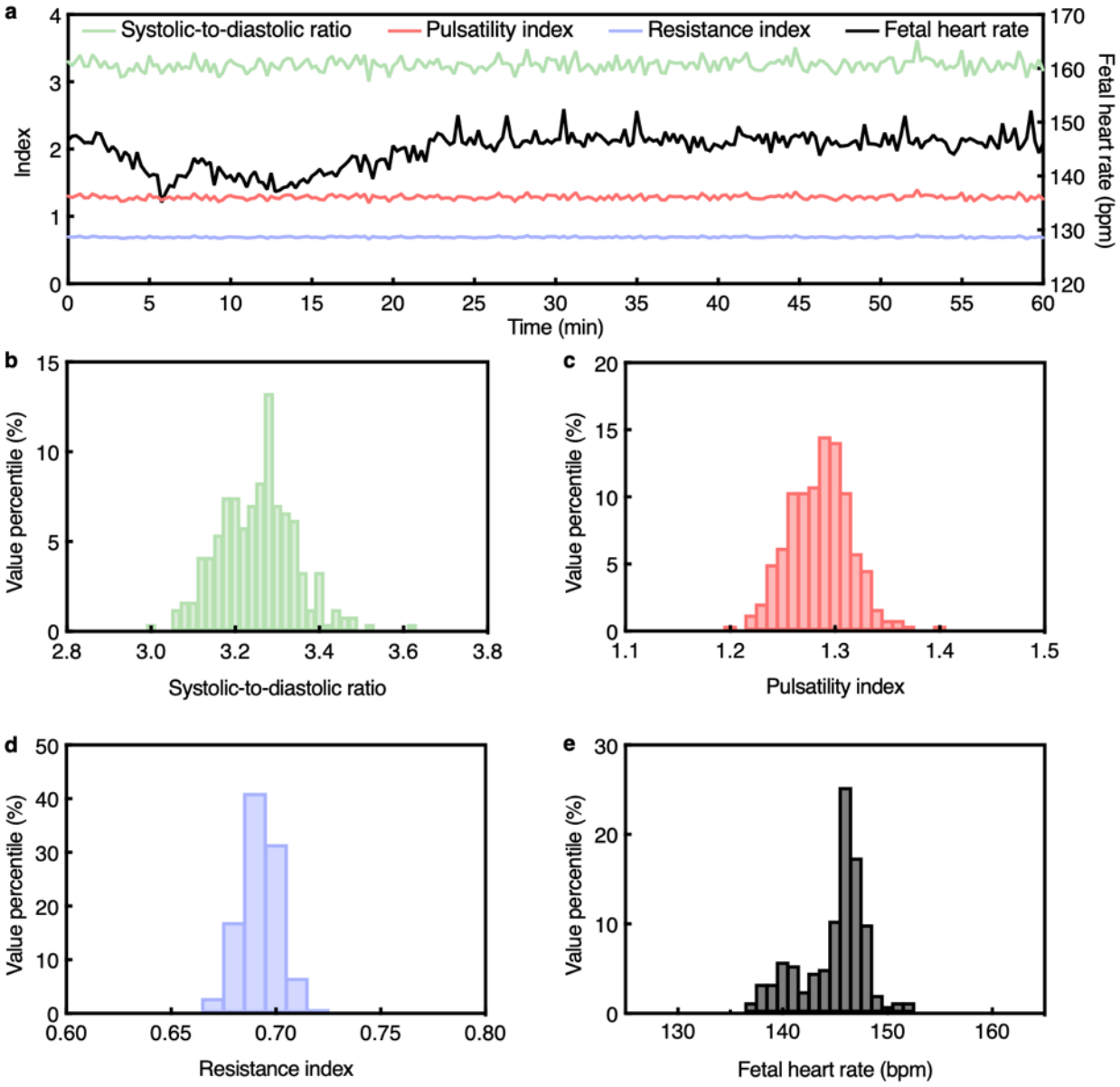
Supplementary Fig. 66 | Continuous fetal monitoring using the UPatch in Participant #31. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



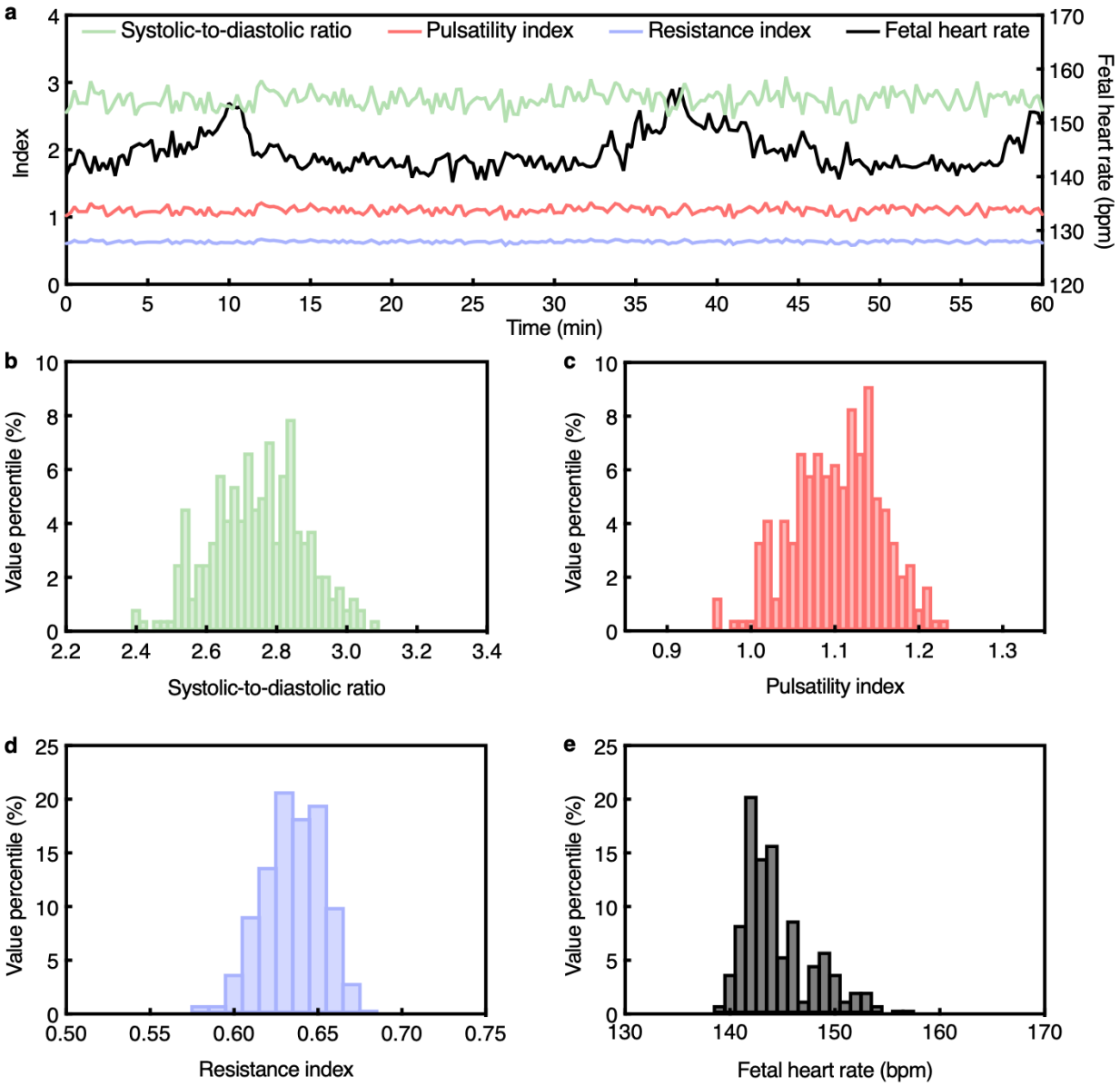
Supplementary Fig. 67 | Continuous fetal monitoring using the UPatch in Participant #32. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



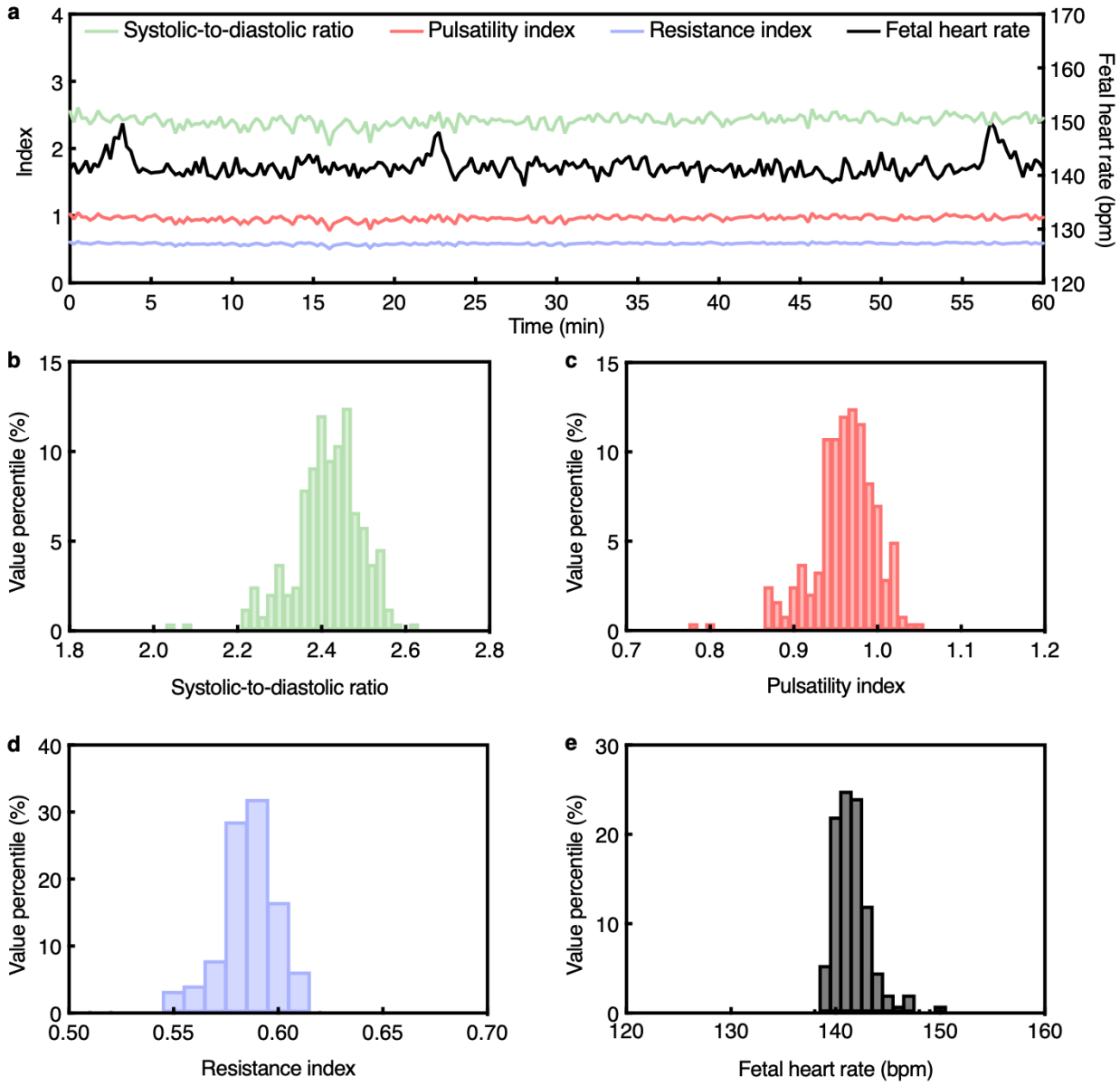
Supplementary Fig. 68 | Continuous fetal monitoring using the UPatch in Participant #33. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



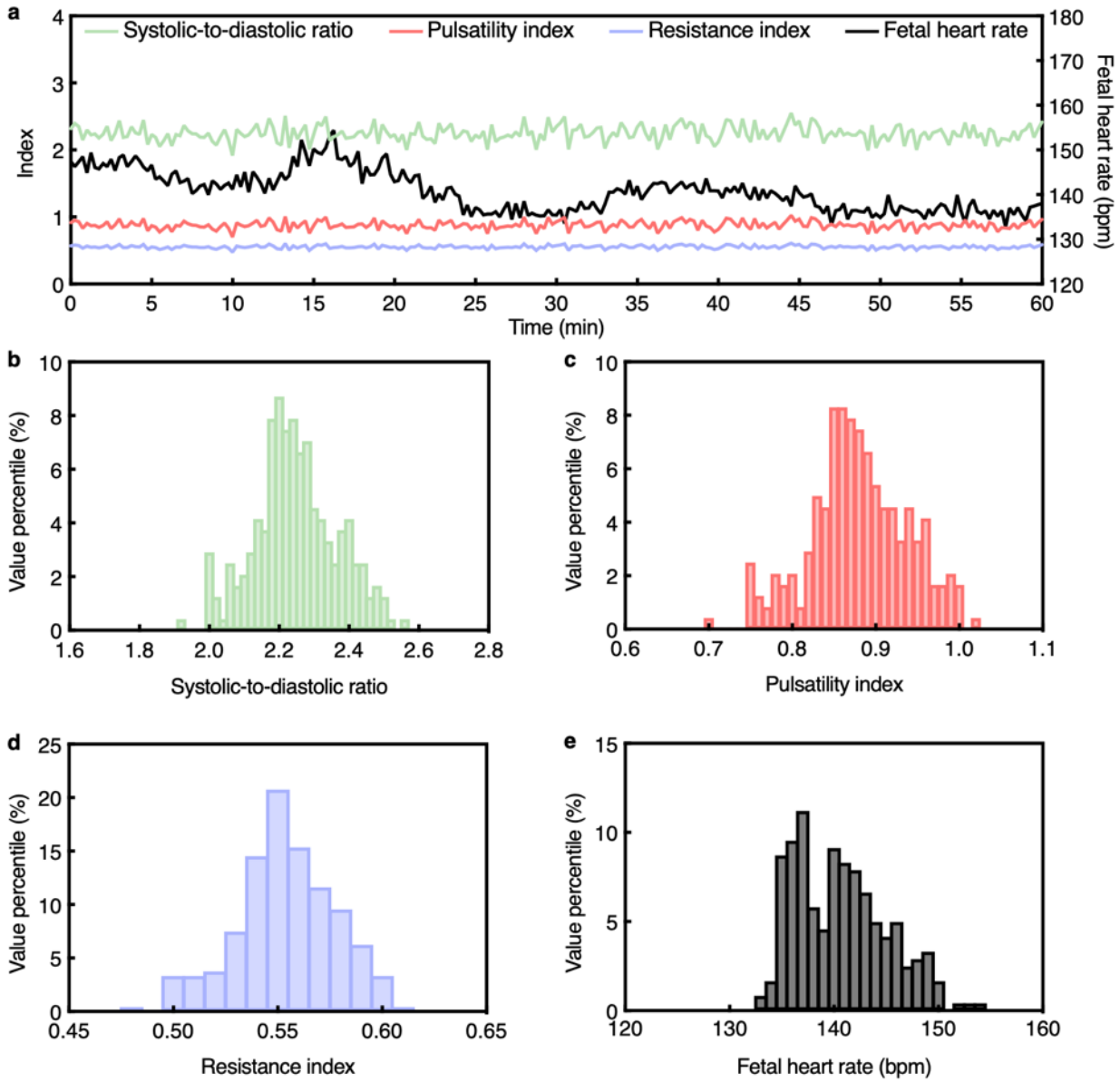
Supplementary Fig. 69 | Continuous fetal monitoring using the UPatch in Participant #34. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



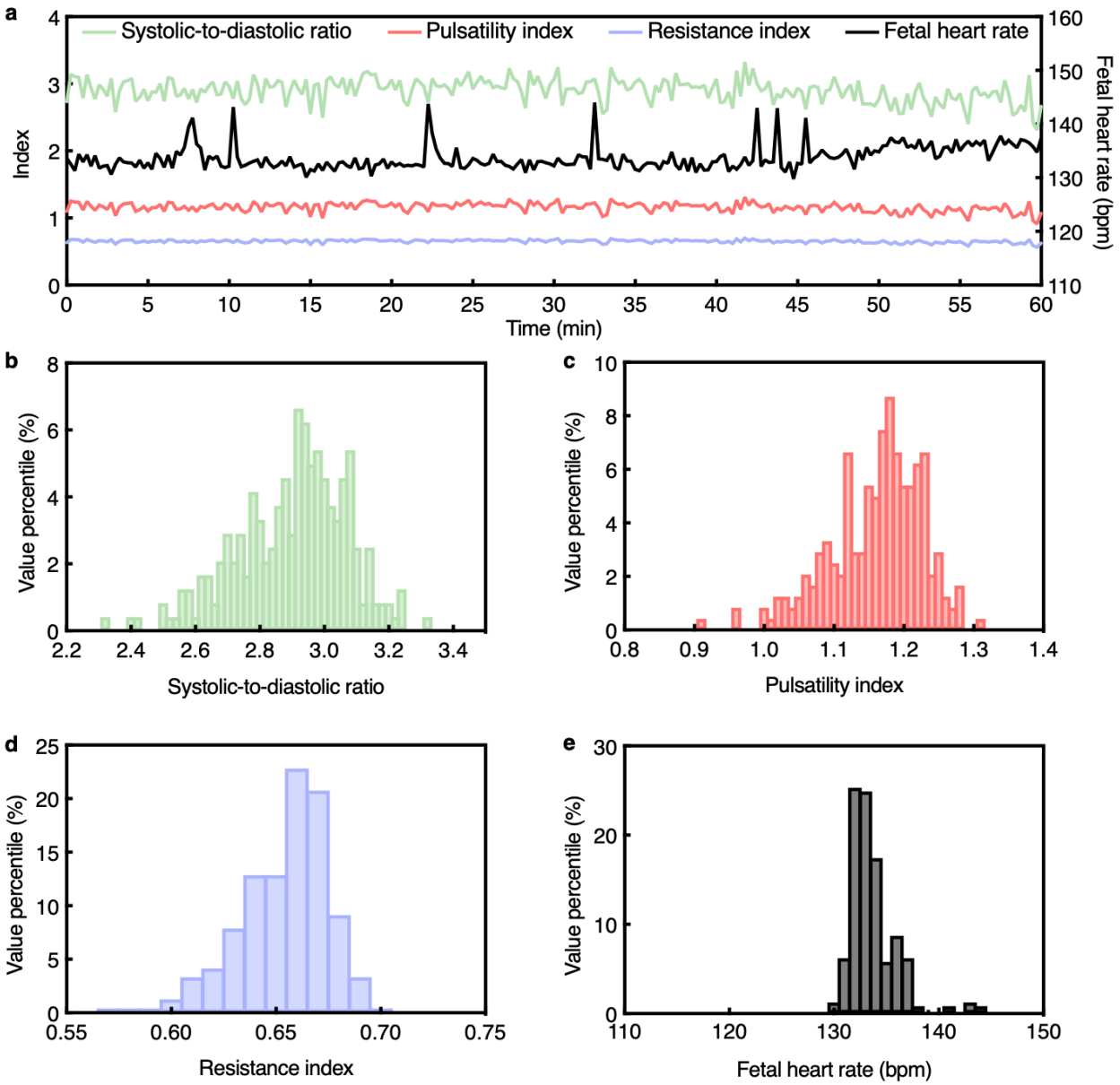
Supplementary Fig. 70 | Continuous fetal monitoring using the UPatch in Participant #35. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



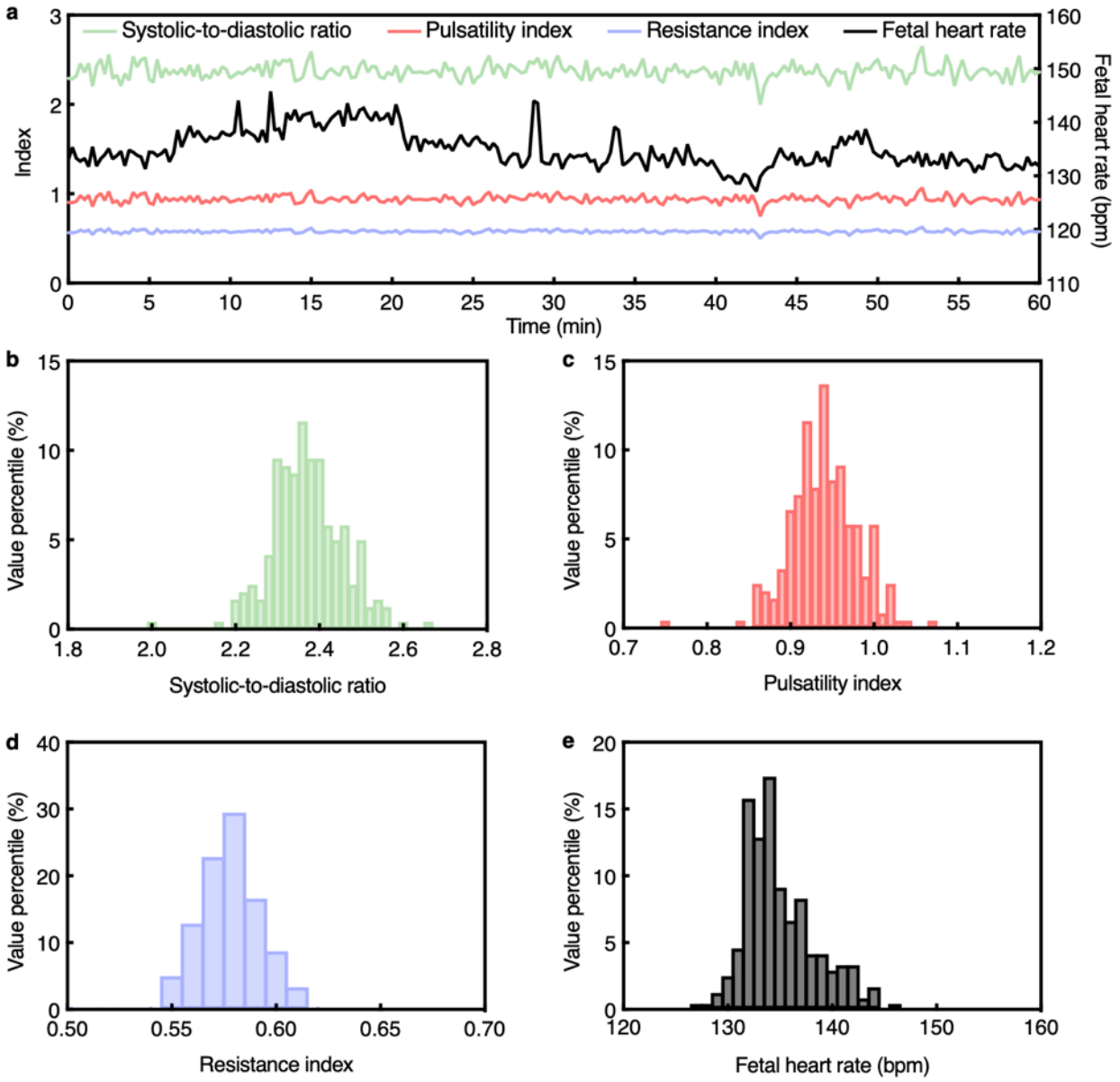
Supplementary Fig. 71 | Continuous fetal monitoring using the UPatch in Participant #36. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



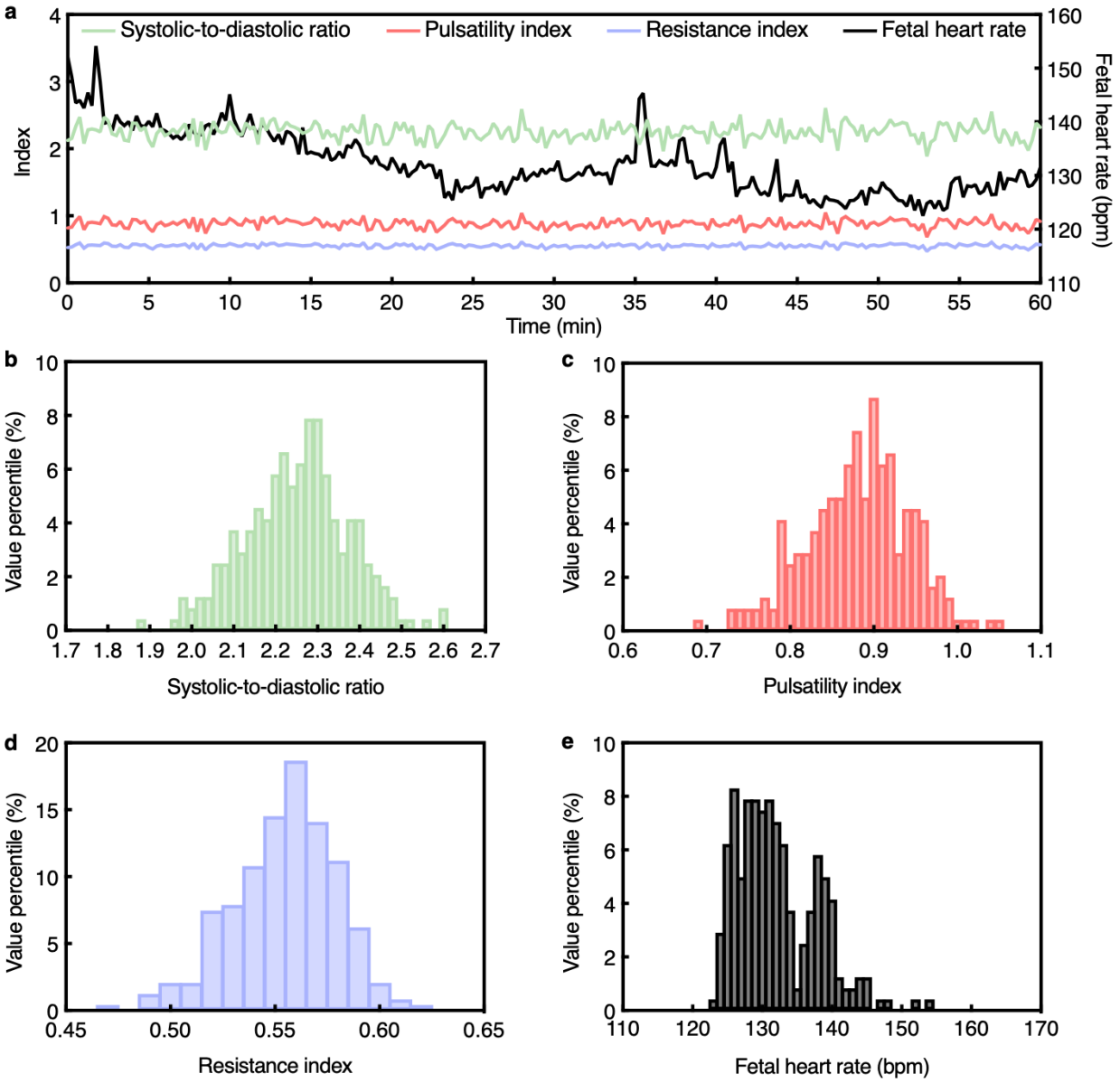
Supplementary Fig. 72 | Continuous fetal monitoring using the UPatch in Participant #37. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



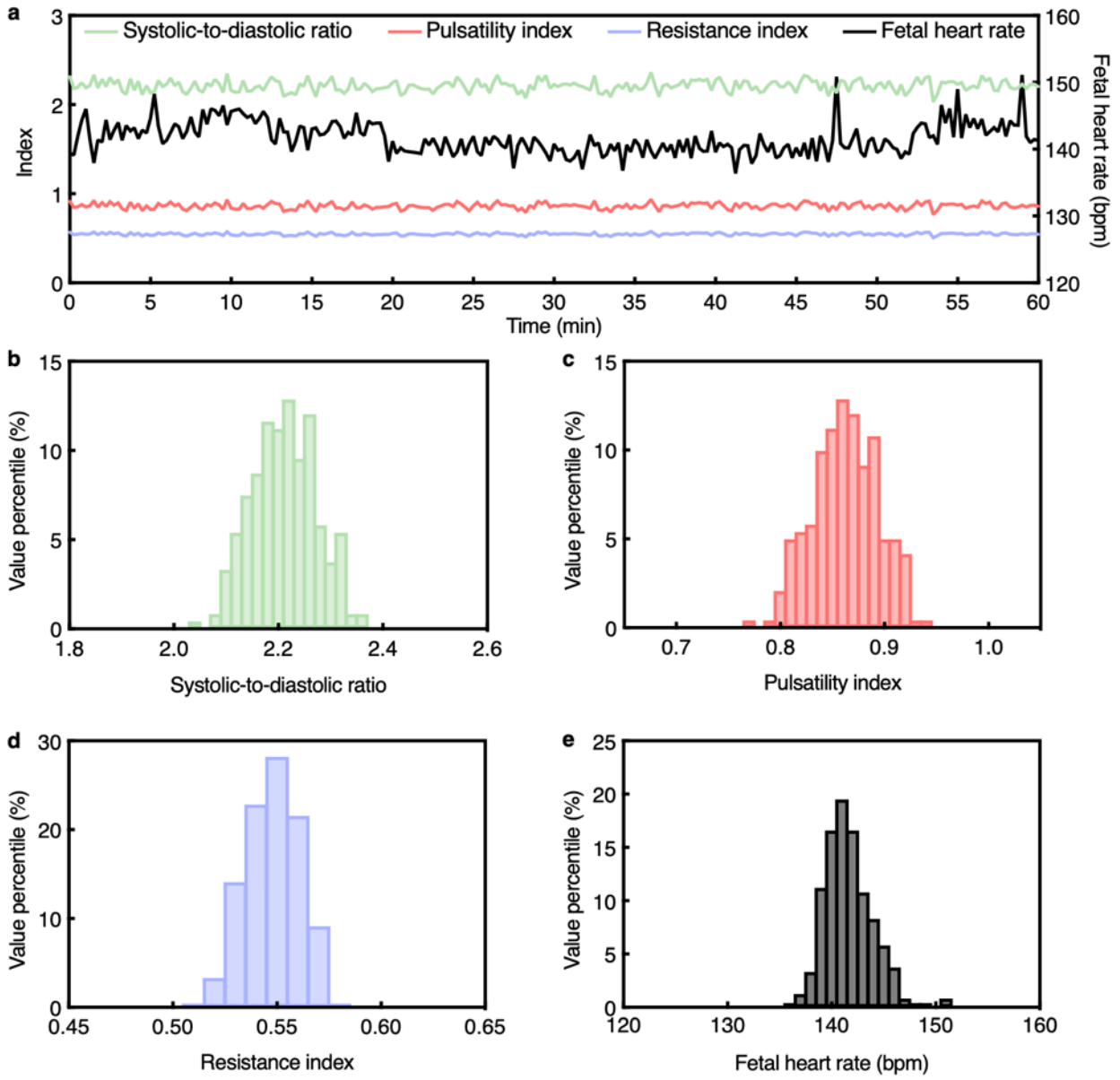
Supplementary Fig. 73 | Continuous fetal monitoring using the UPatch in Participant #38. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



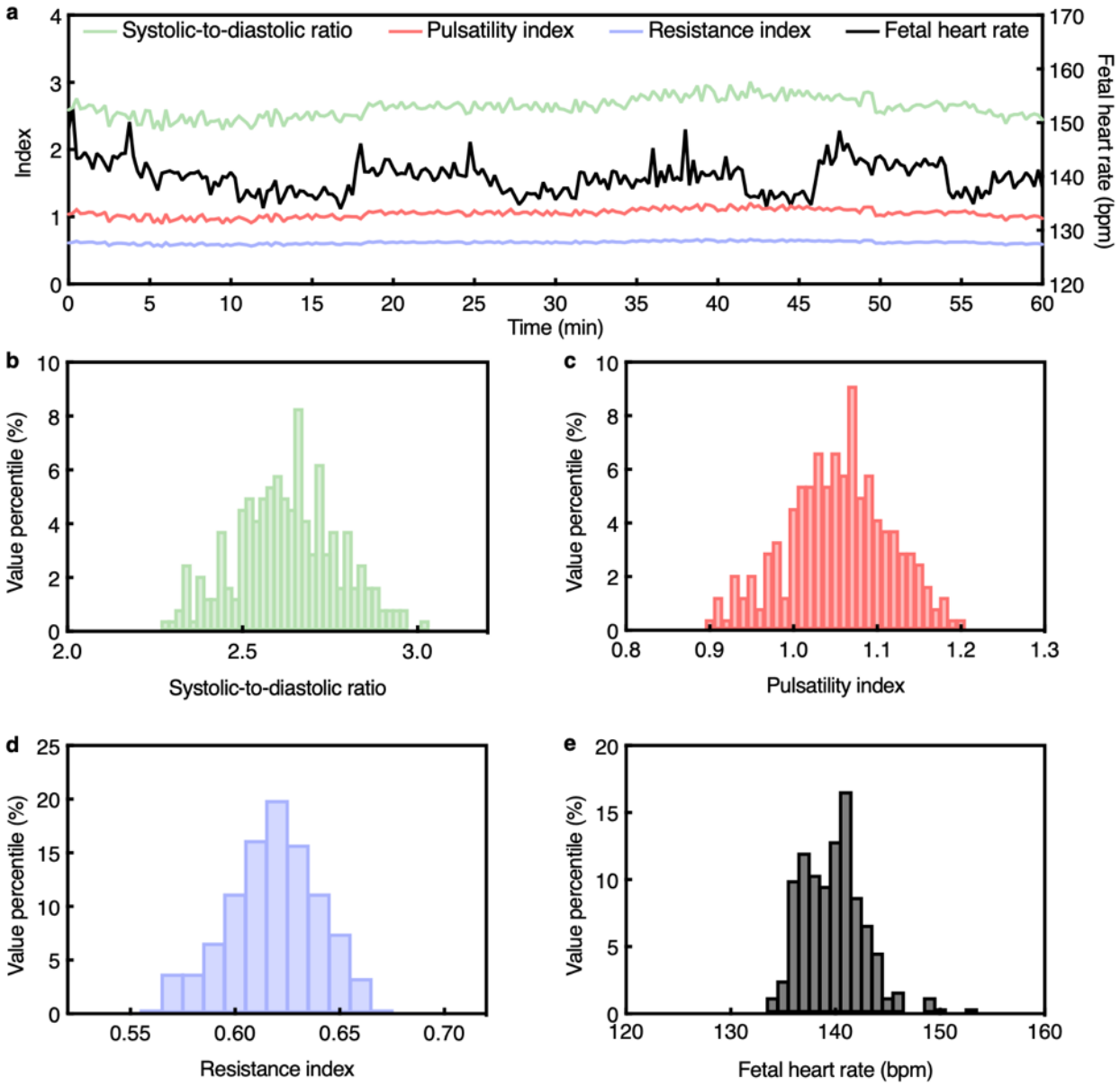
Supplementary Fig. 74 | Continuous fetal monitoring using the UPatch in Participant #39. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



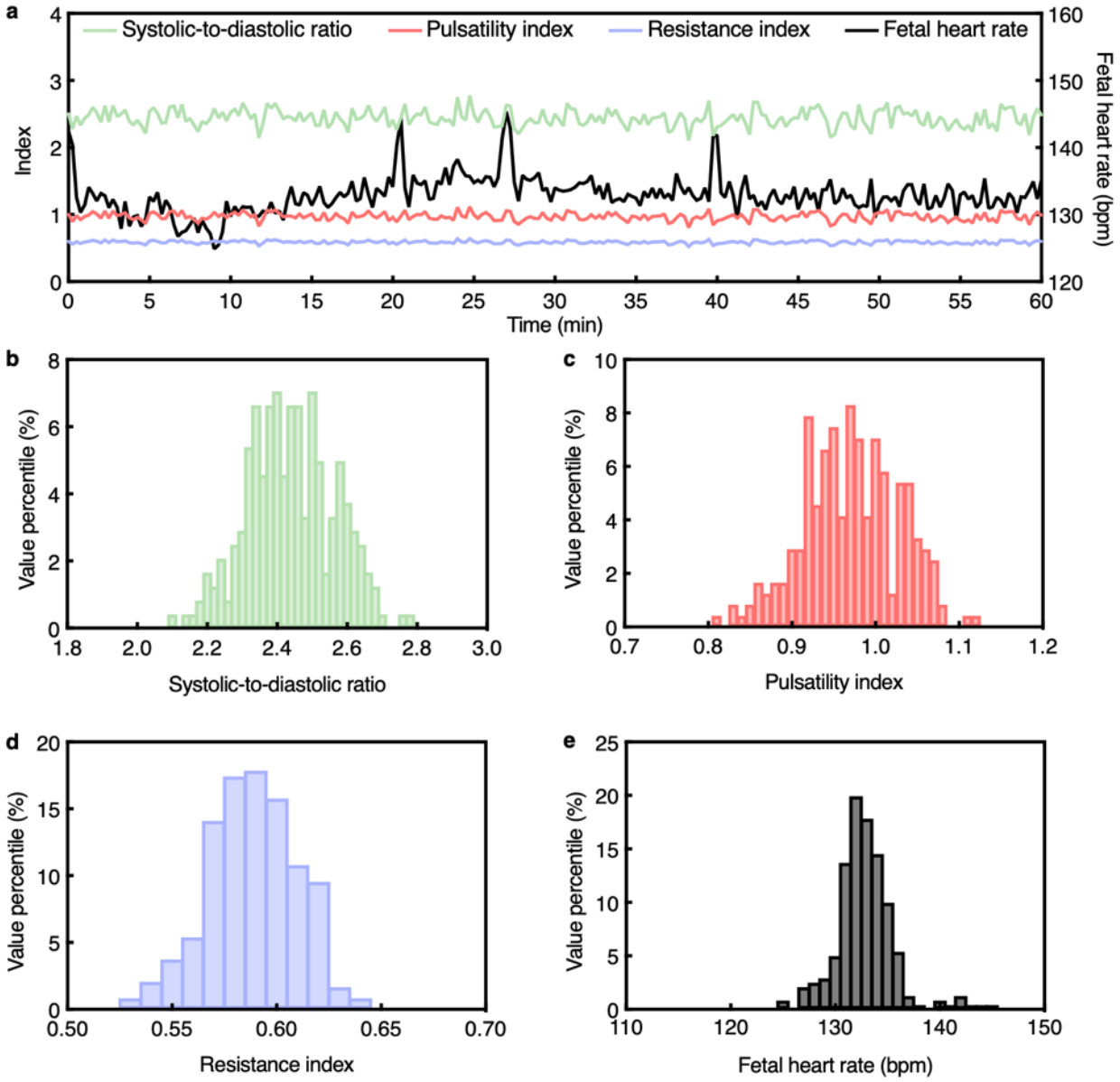
Supplementary Fig. 75 | Continuous fetal monitoring using the UPatch in Participant #40. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



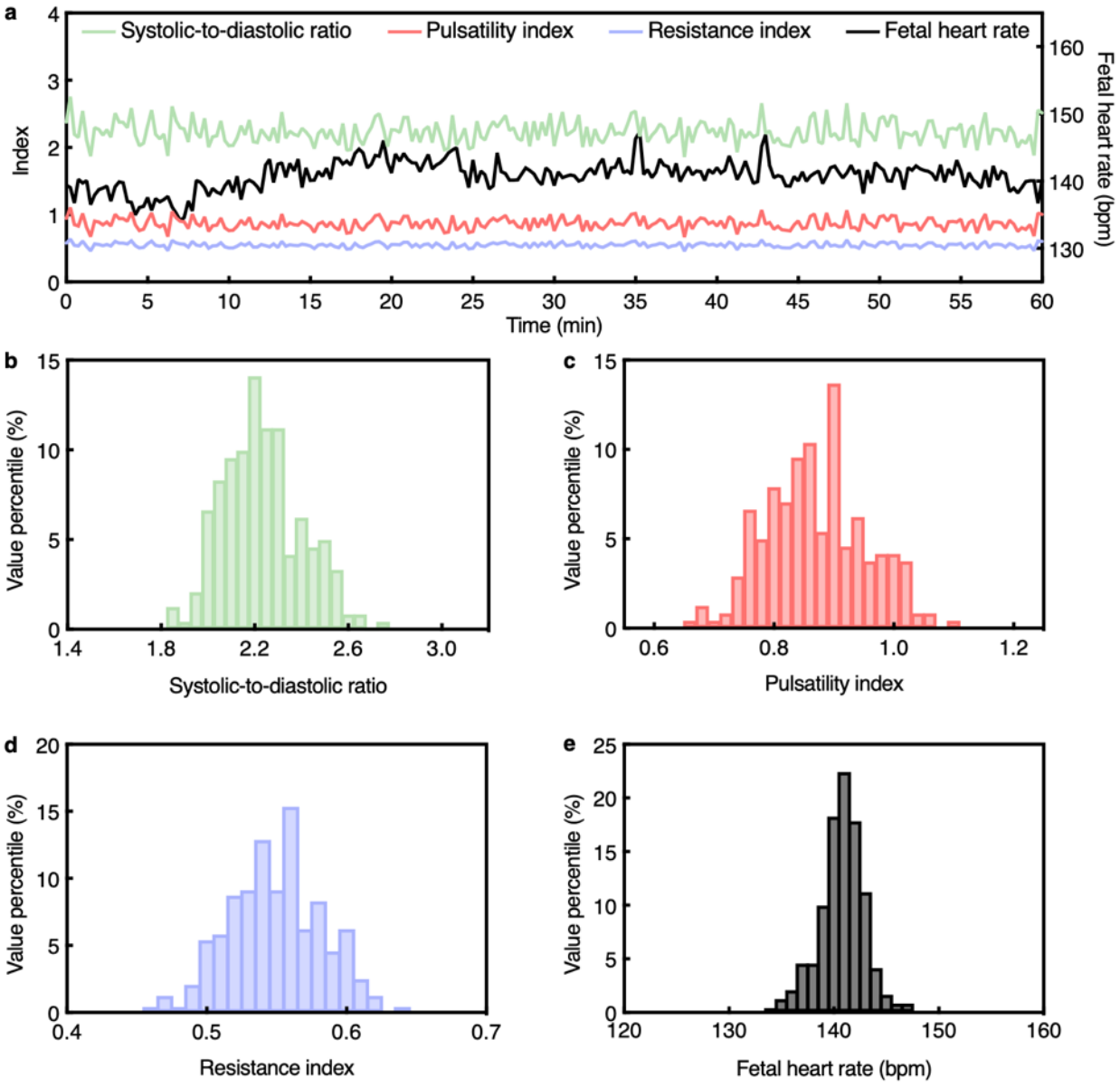
Supplementary Fig. 76 | Continuous fetal monitoring using the UPatch in Participant #41. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



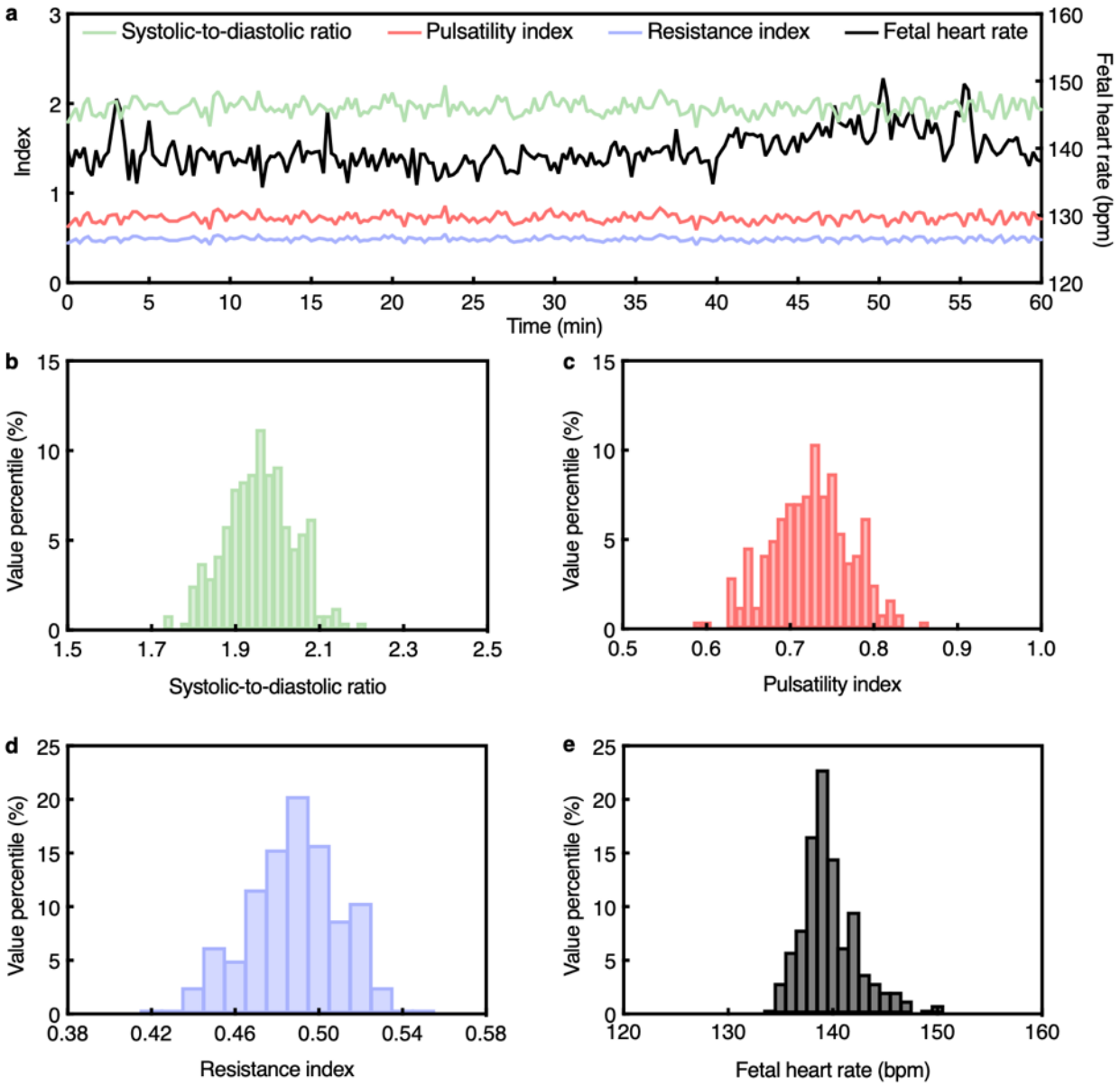
Supplementary Fig. 77 | Continuous fetal monitoring using the UPatch in Participant #42. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



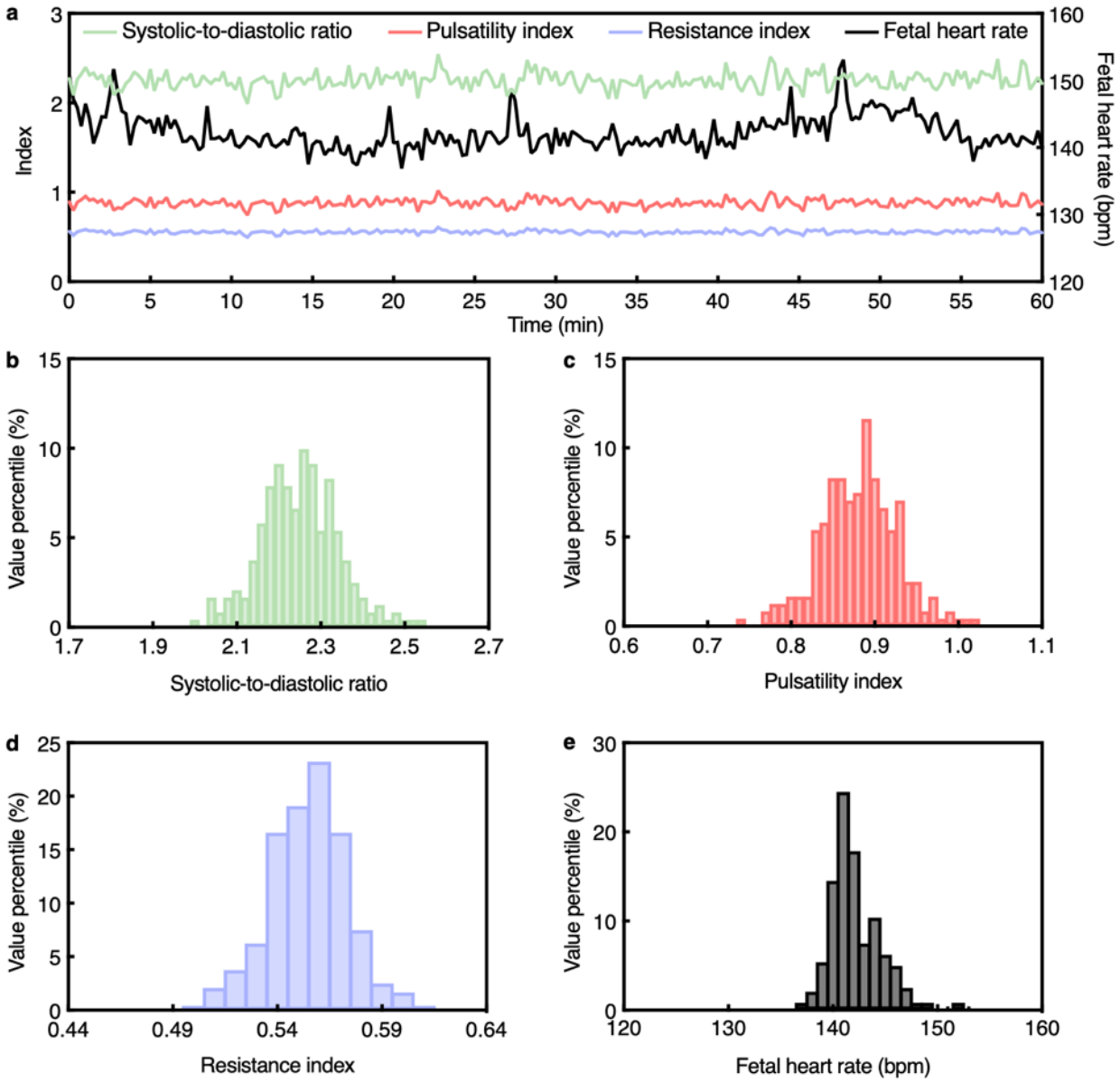
Supplementary Fig. 78 | Continuous fetal monitoring using the UPatch in Participant #43. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



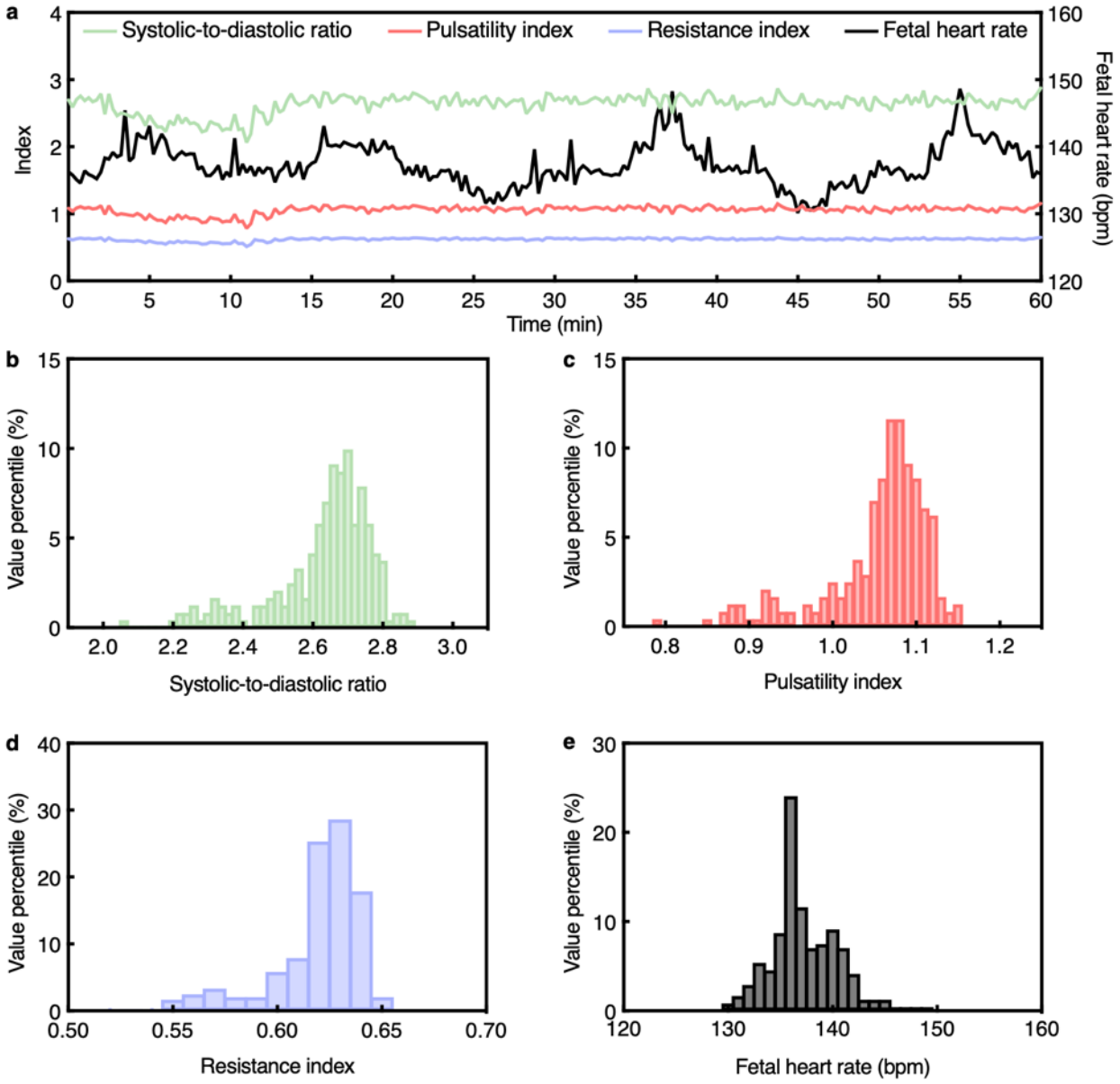
Supplementary Fig. 79 | Continuous fetal monitoring using the UPatch in Participant #44. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



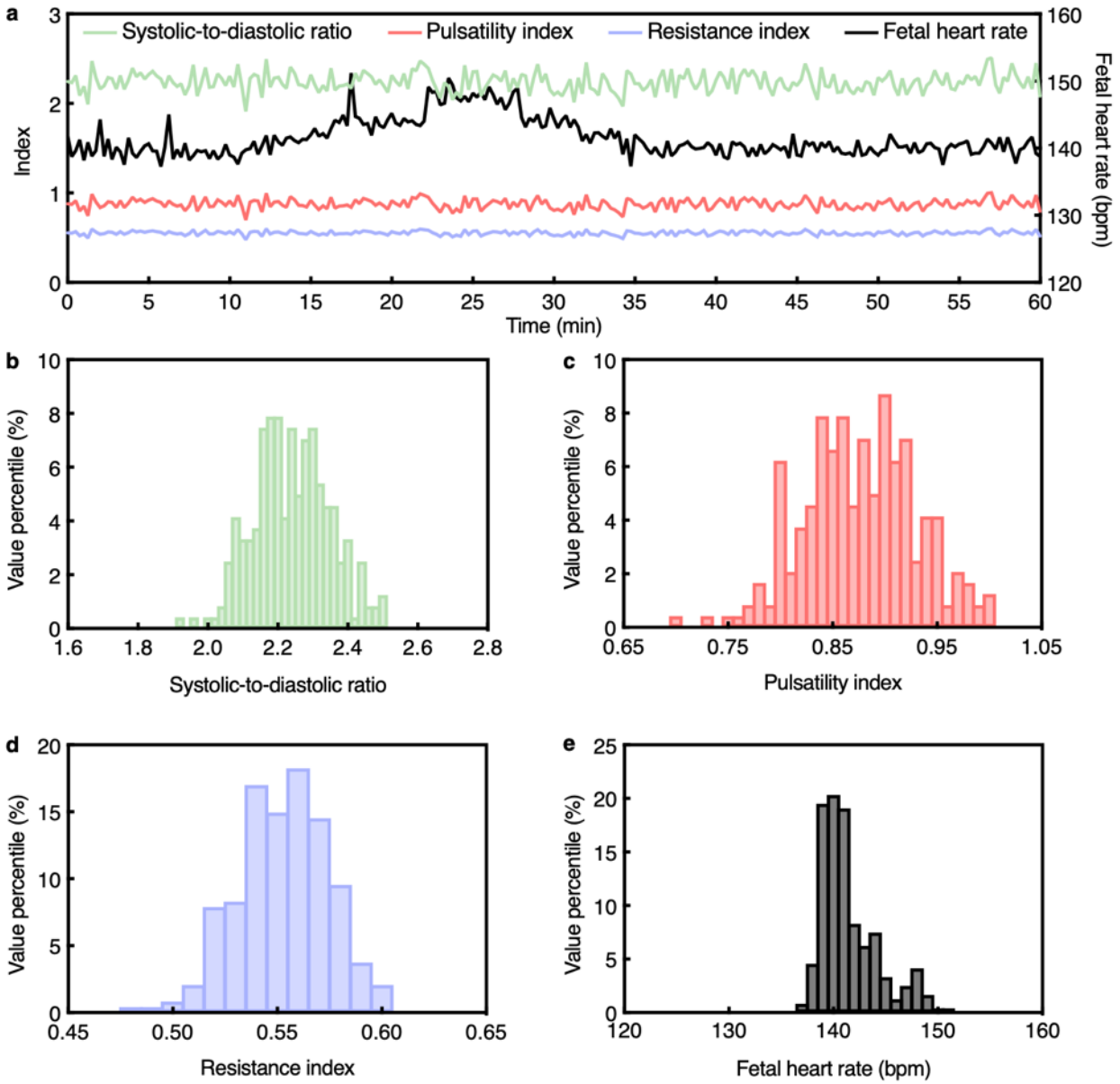
Supplementary Fig. 80 | Continuous fetal monitoring using the UPatch in Participant #45. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



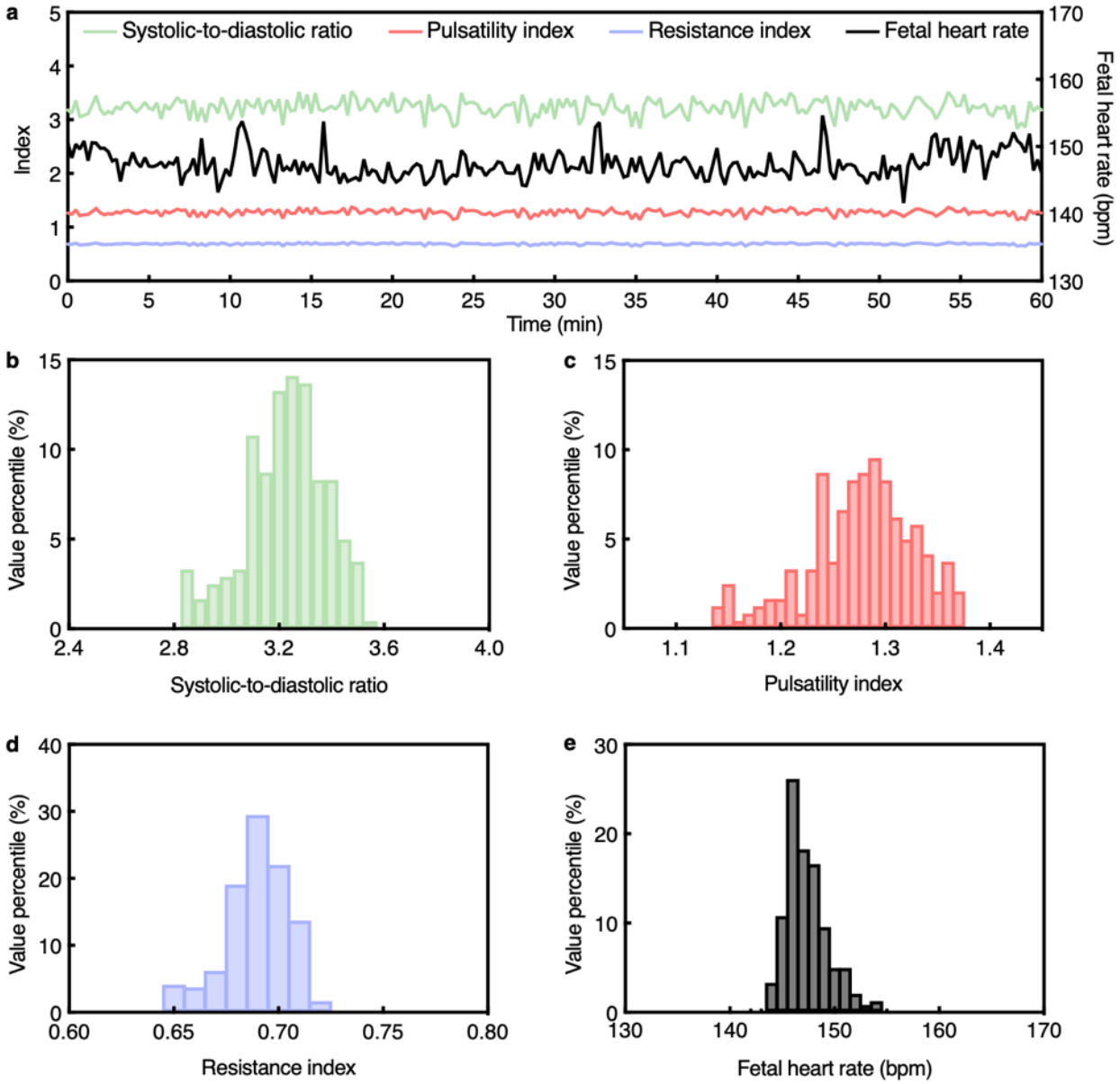
Supplementary Fig. 81 | Continuous fetal monitoring using the UPatch in Participant #46. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



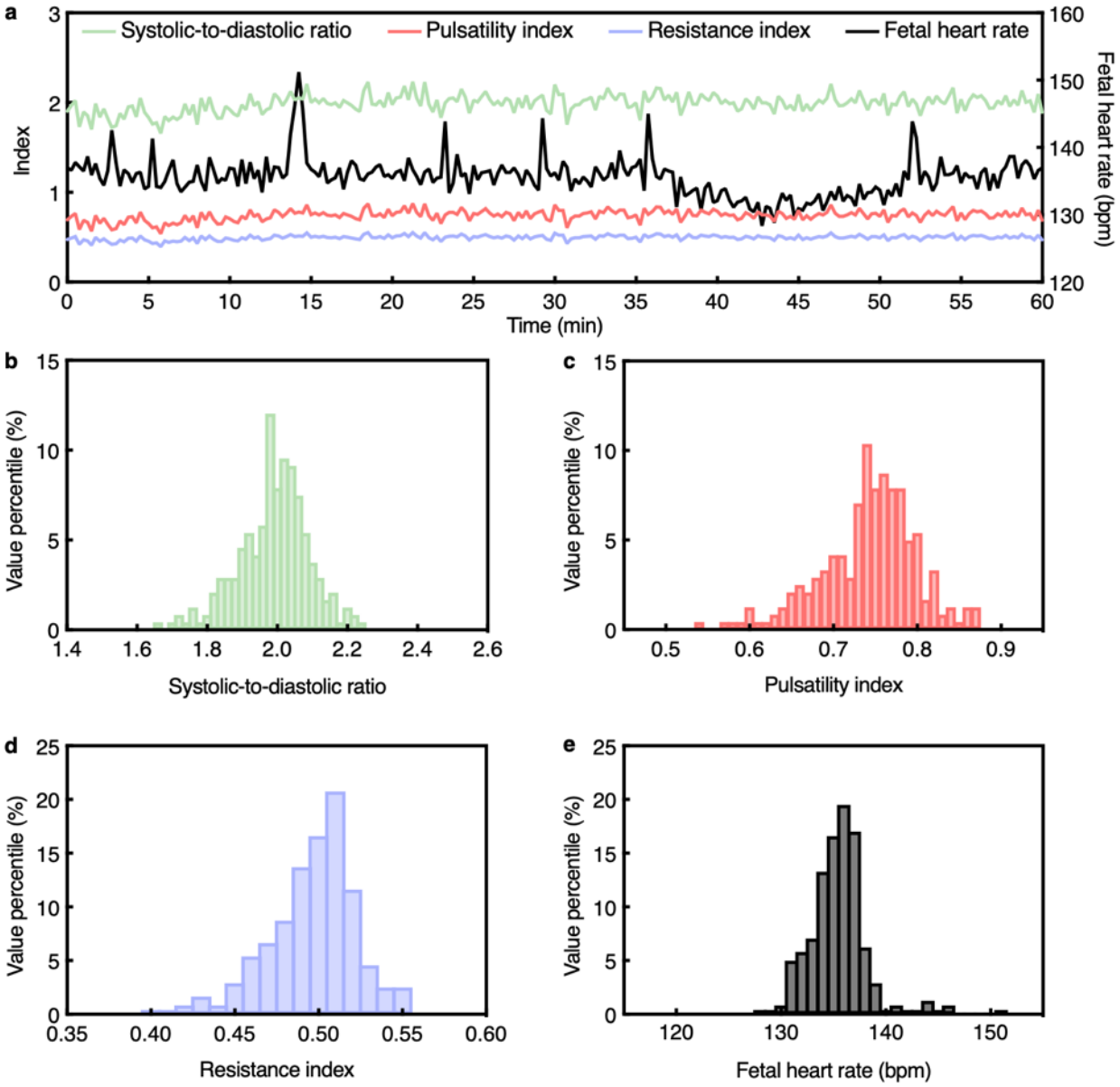
Supplementary Fig. 82 | Continuous fetal monitoring using the UPatch in Participant #47. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



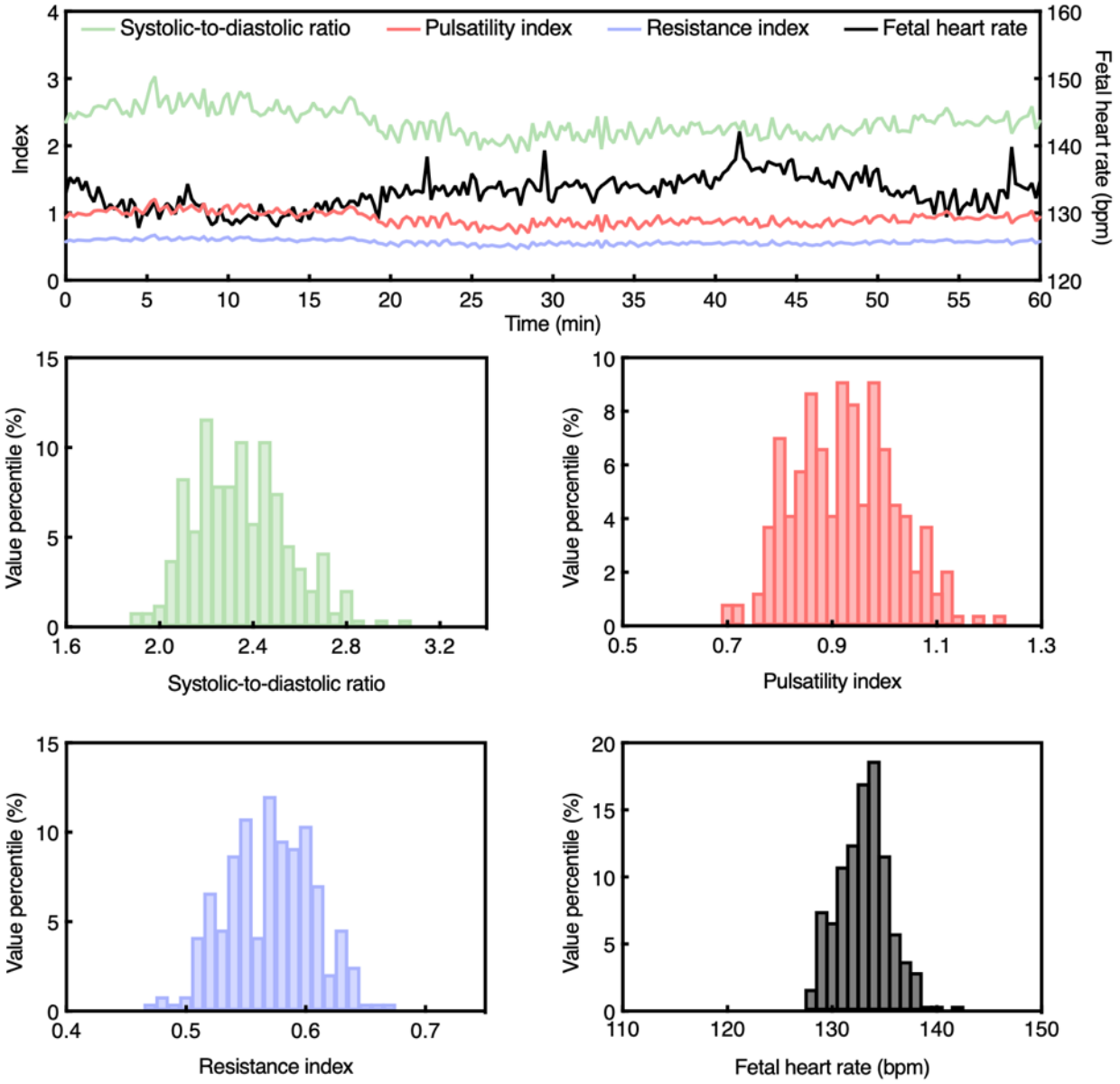
Supplementary Fig. 83 | Continuous fetal monitoring using the UPatch in Participant #48. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



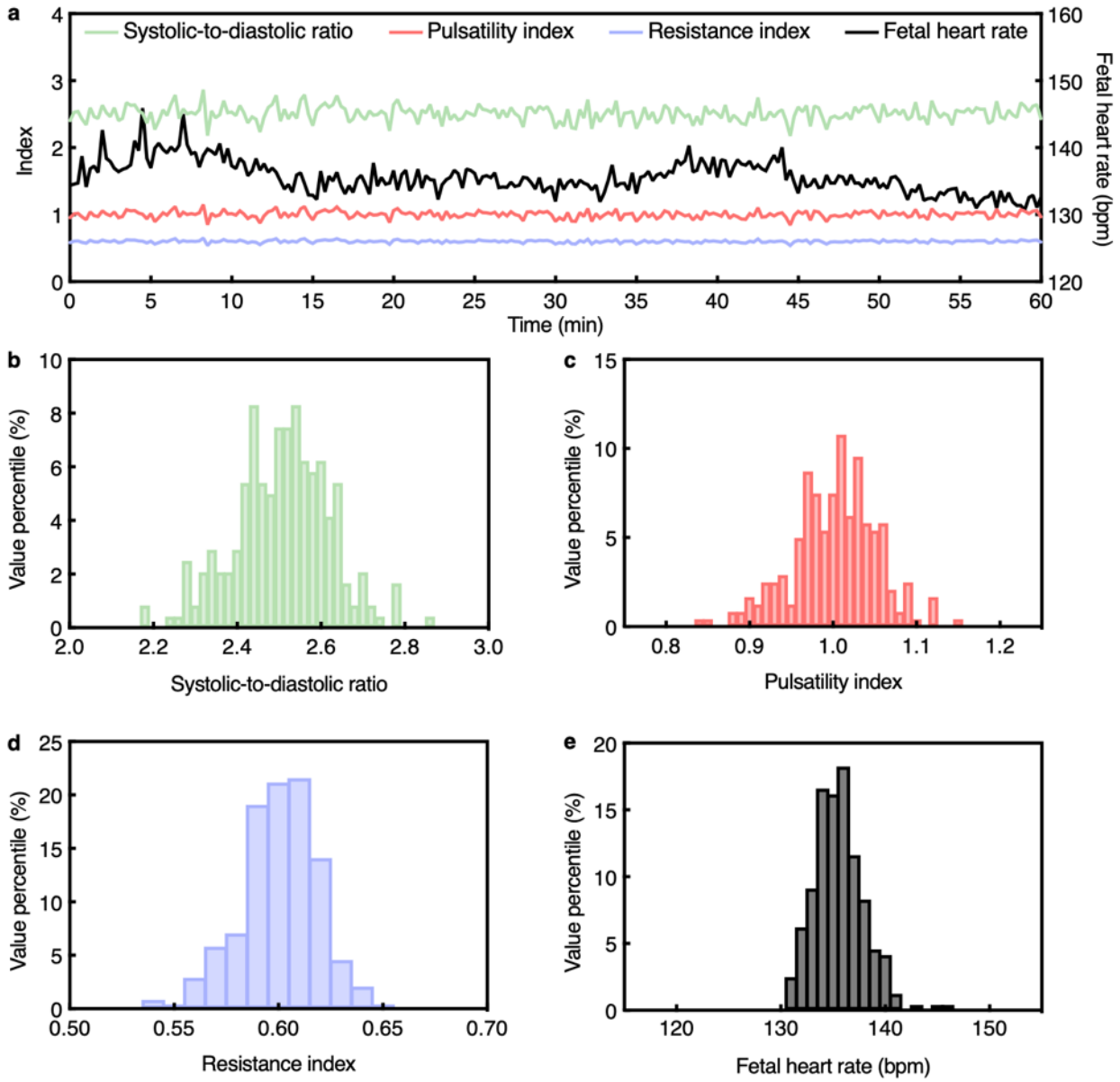
Supplementary Fig. 84 | Continuous fetal monitoring using the UPatch in Participant #49. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



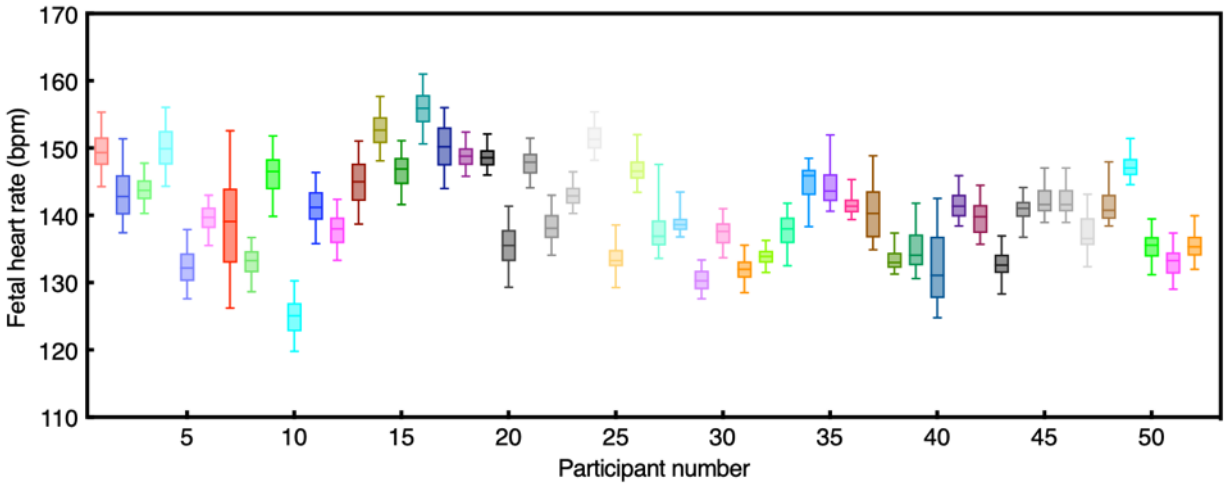
Supplementary Fig. 85 | Continuous fetal monitoring using the UPatch in Participant #50. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



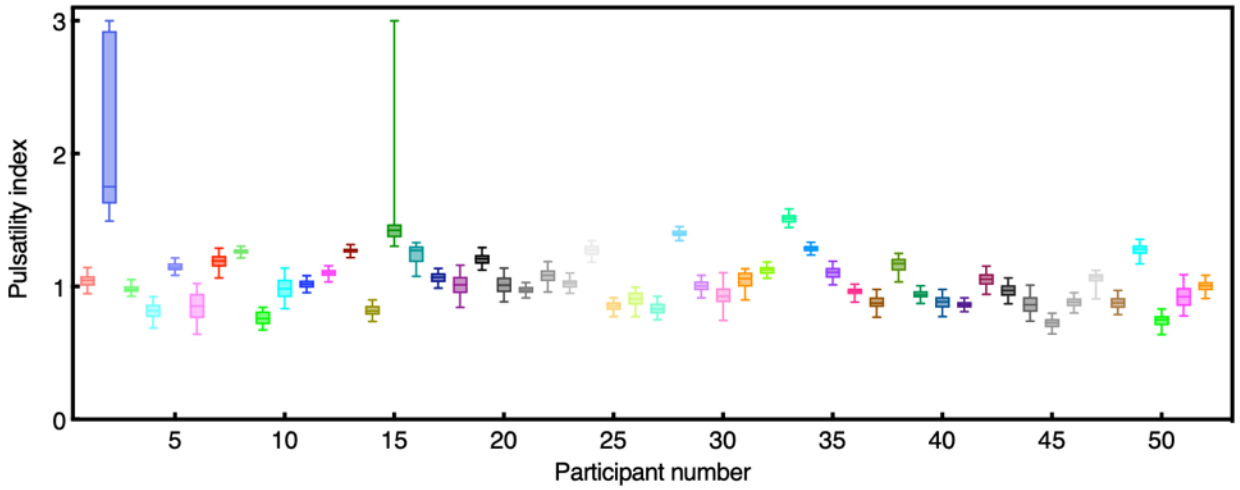
Supplementary Fig. 86 | Continuous fetal monitoring using the UPatch in Participant #51. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



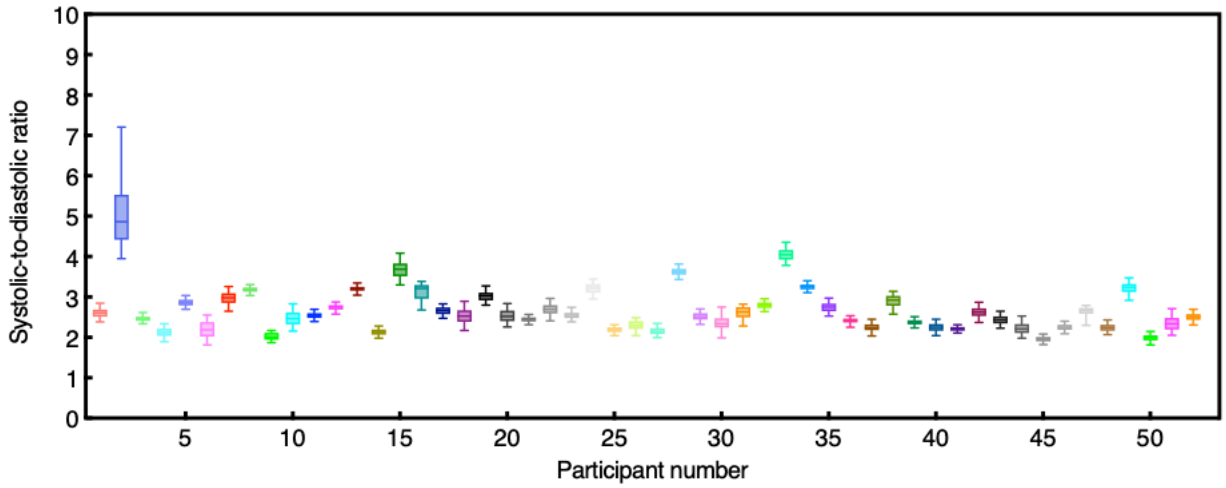
Supplementary Fig. 87 | Continuous fetal monitoring using the UPatch in Participant #52. a, Time series data from the 1-h continuous monitoring session. Systolic-to-diastolic ratio, pulsatility index, and resistance index are plotted on the left y-axis, and fetal heart rate is plotted on the right y-axis. Histograms of **b**, systolic-to-diastolic ratio, **c**, pulsatility index, **d**, resistance index, and **e**, fetal heart rate over the recording period.



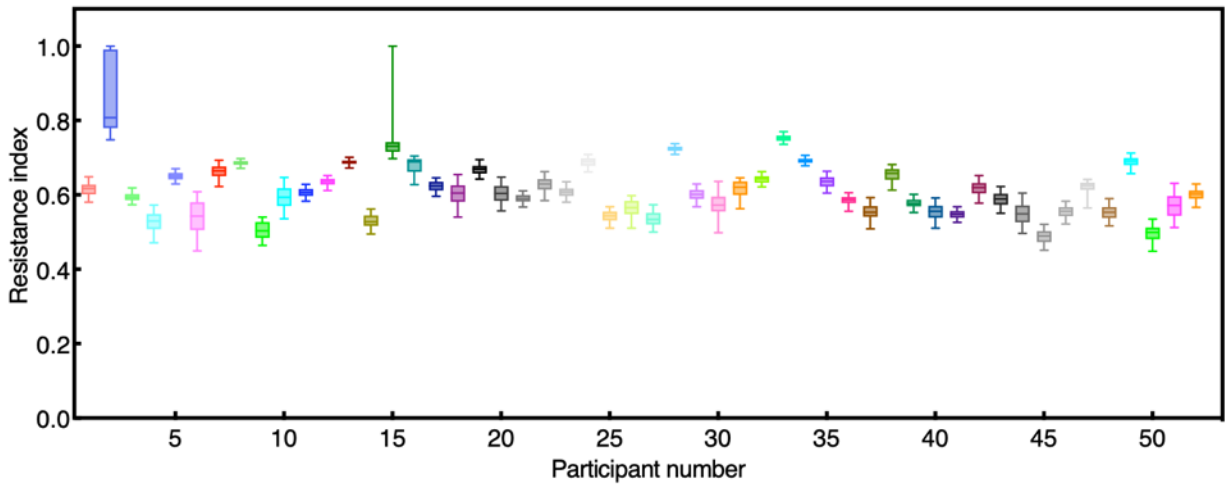
Supplementary Fig. 88 | Box plots of fetal heart rate during continuous monitoring. Distribution of fetal heart rate values across the full monitoring session for all participants. Each box represents the interquartile range (25th to 75th percentiles), the whiskers denote the 5th to 95th percentile range, and the midline indicates the median. The observed variability reflects dynamic physiological changes during the monitoring period.



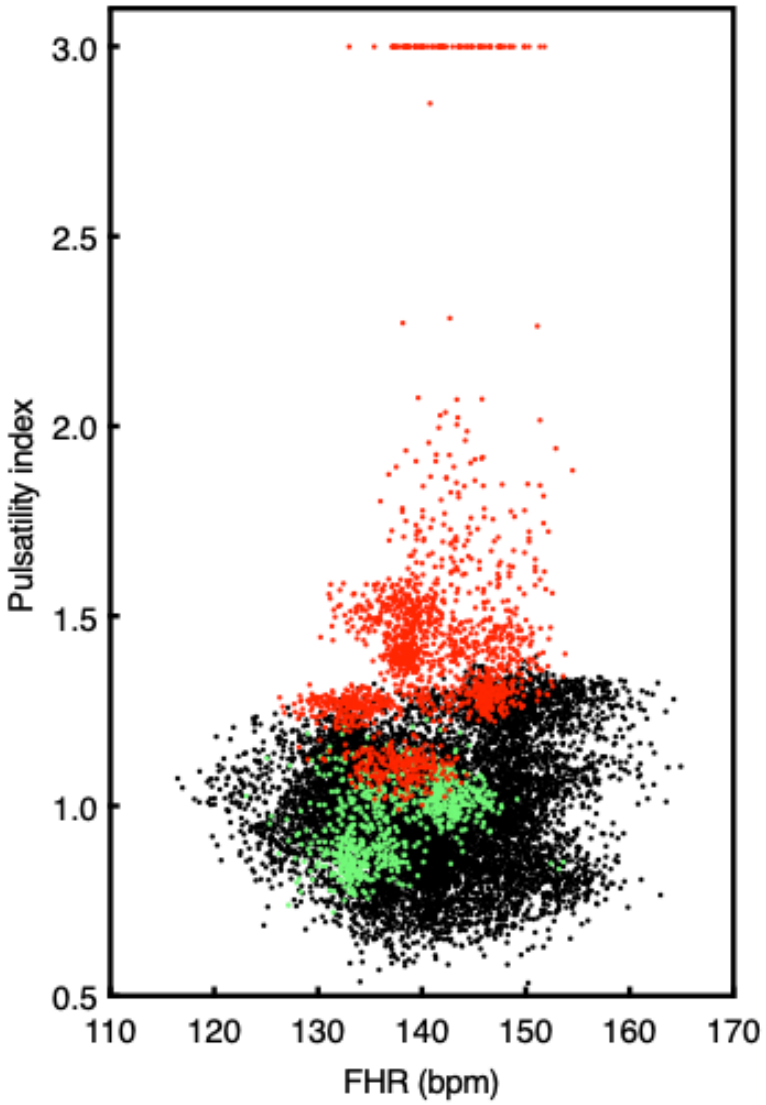
Supplementary Fig. 89 | Box plots of pulsatility index during continuous monitoring. Distribution of pulsatility index values across the full monitoring session for all participants. Each box represents the interquartile range (25th to 75th percentiles), the whiskers denote the 5th to 95th percentile range, and the midline indicates the median. The observed variability reflects dynamic physiological changes during the monitoring period.



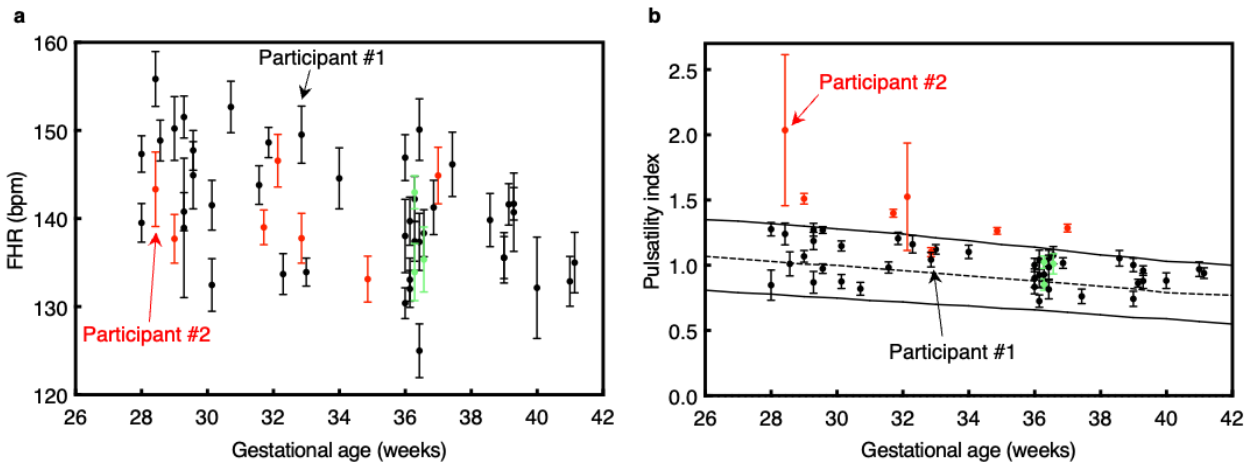
Supplementary Fig. 90 | Box plots of systolic-to-diastolic ratio during continuous monitoring. Distribution of systolic-to-diastolic ratio values across the full monitoring session for all participants. Each box represents the interquartile range (25th to 75th percentiles), the whiskers denote the 5th to 95th percentile range, and the midline indicates the median. The observed variability reflects dynamic physiological changes during the monitoring period. In Participants #2 and #15, systolic-to-diastolic ratios that could not be computed due to absent end diastolic flow were excluded from the analysis; only computable values were included.



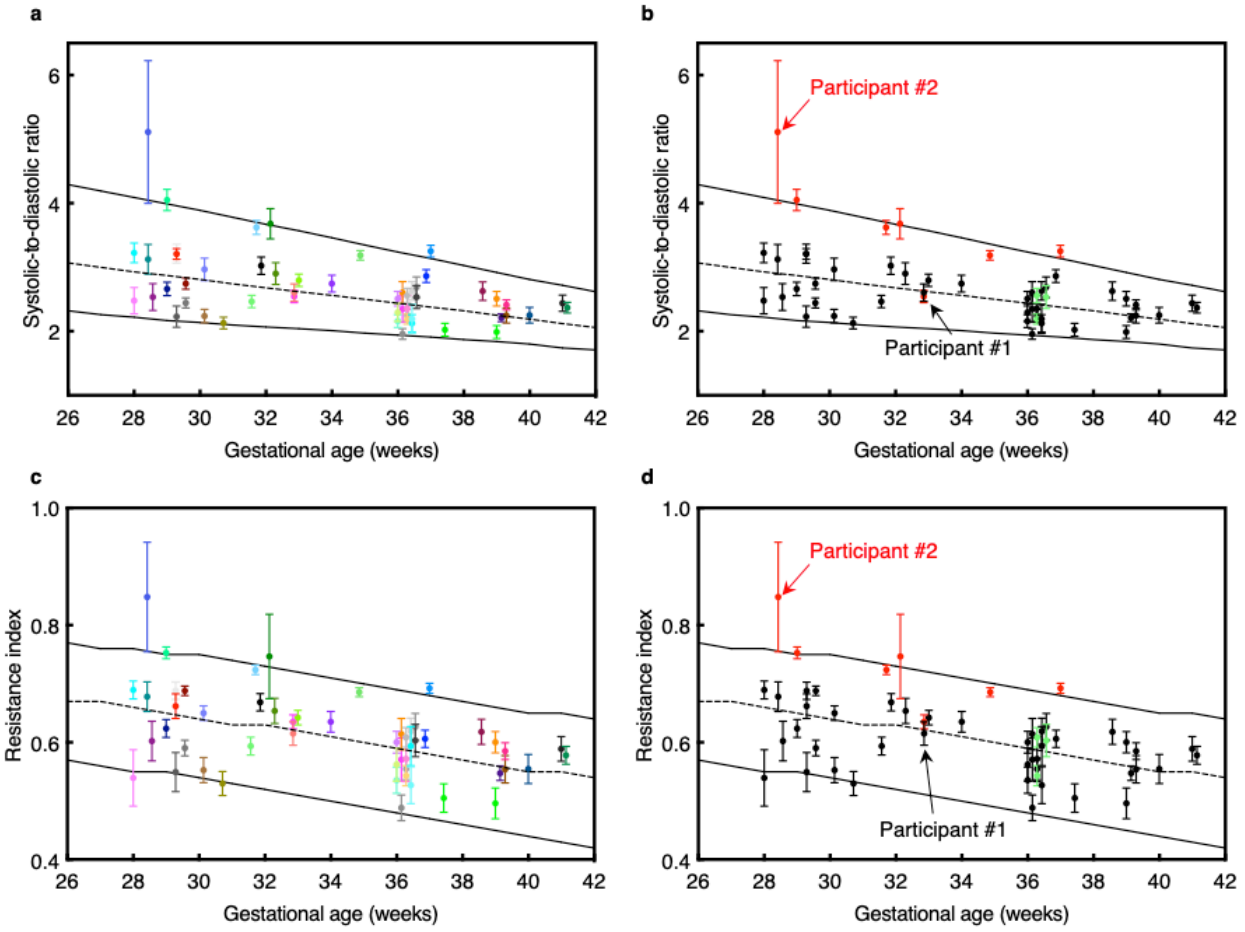
Supplementary Fig. 91 | Box plots of resistance index during continuous monitoring. Distribution of resistance index values across the full monitoring session for all participants. Each box represents the interquartile range (25th to 75th percentiles), the whiskers denote the 5th to 95th percentile range, and the midline indicates the median. The observed variability reflects dynamic physiological changes during the monitoring period.



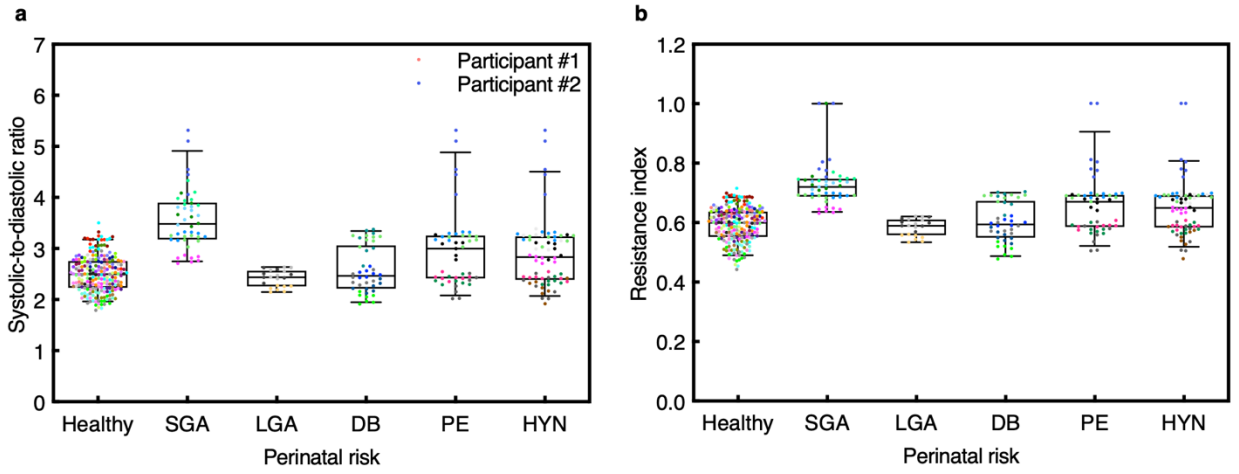
Supplementary Fig. 92 | Scatterplot of pulsatility index against FHR. The color corresponds to perinatal conditions: healthy (black, 42 participants), small for gestational age (red, 7 participants), and large for gestational age (green, 3 participants). Because participants are often presented with multiple co-occurring maternal risk conditions, risk groups were classified solely based on fetal risk conditions.



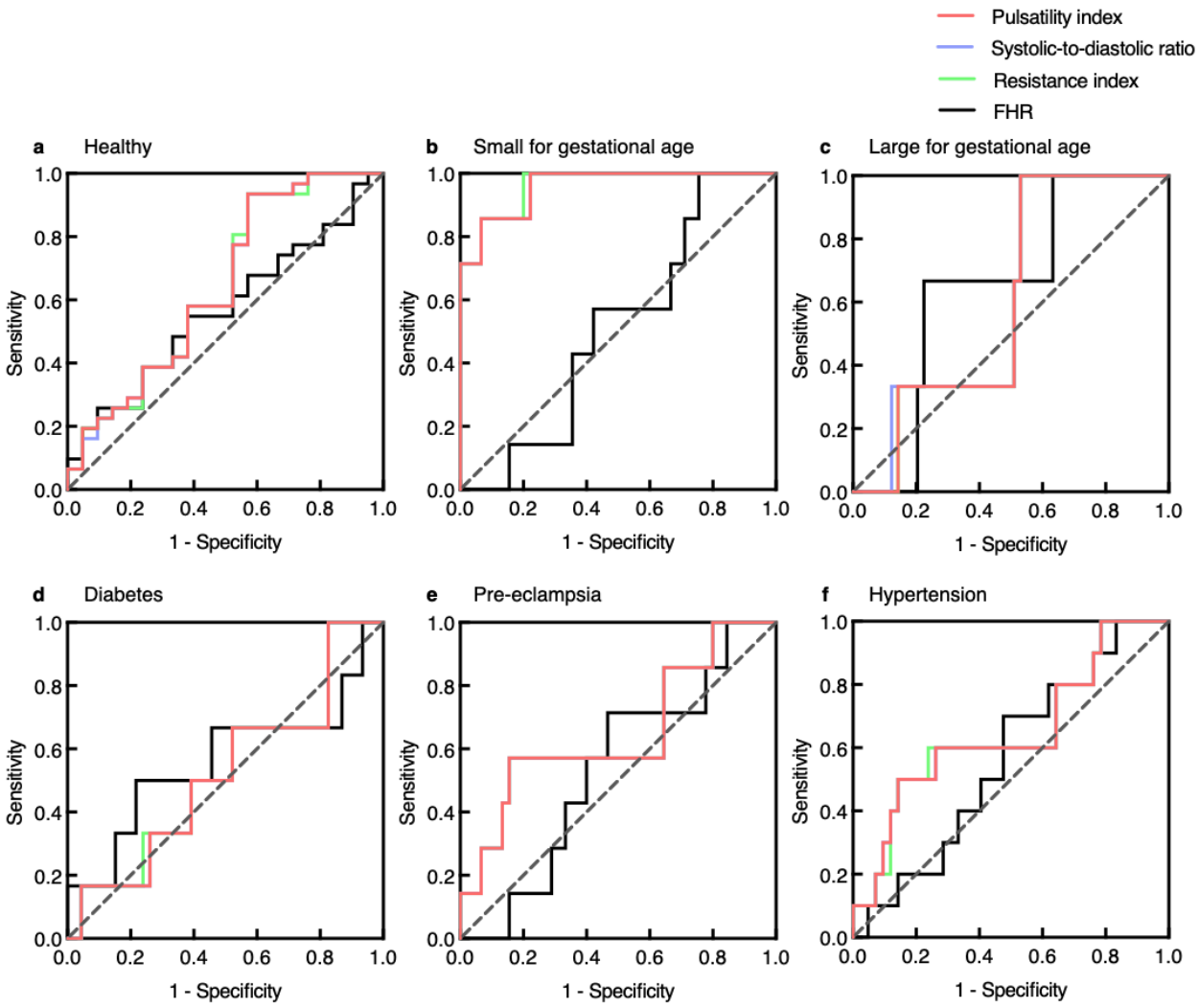
Supplementary Fig. 93 | Gestational trends in FHR and pulsatility index. a, FHR plotted against gestational age. **b**, Umbilical artery pulsatility index plotted against gestational age. The color corresponds to perinatal conditions: healthy (black, 42 participants), small for gestational age (red, 7 participants), and large for gestational age (green, 3 participants). The black dashed line is the 50th percentile, and the solid black lines are the 5th and 95th percentiles of a widely used reference population[68]. Because participants are often presented with multiple co-occurring maternal risk conditions, risk groups were classified solely based on fetal risk conditions. Data are presented as mean ± standard deviation (n = 241).



Supplementary Fig. 94 | Gestational trends in systolic-to-diastolic ratio and resistance index. **a**,**b**, Umbilical artery systolic-to-diastolic ratio plotted against gestational age (**a**) with each color corresponding to an individual participant and (**b**) with each color corresponding to perinatal conditions: healthy (black, 42 participants), small for gestational age (red, 7 participants), and large for gestational age (green, 3 participants). **c**,**d**, Umbilical artery resistance index plotted against gestational age (**c**) with each color corresponding to an individual participant and (**d**) with each color corresponding to perinatal conditions: healthy (black, 42 participants), small for gestational age (red, 7 participants), and large for gestational age (green, 3 participants). The black dashed line in each plot is the 50th percentile, and the solid black lines are the 5th and 95th percentiles of a widely used reference population[68]. Because participants are often presented with multiple co-occurring maternal risk conditions, risk groups were classified solely based on fetal risk conditions. Data are presented as mean ± standard deviations (n = 241).



Supplementary Fig. 95 | Risk stratification in systolic-to-diastolic ratio and resistance index. (**a**) Box plot of systolic-to-diastolic ratio stratified by perinatal conditions: healthy (31 participants, 217 data points), small for gestational age (SGA; 7 participants, 46 data points), large for gestational age (LGA; 3 participants, 21 data points), diabetes (DB; 6 participants, 42 data points), pre-eclampsia (PE; 7 participants, 47 data points), and maternal hypertension (HYN; 10 participants, 68 data points). In Participants #2 and #15, systolic-to-diastolic ratios that could not be computed due to absent end diastolic flow were excluded from the analysis; only computable values were included. (**b**) Box plot of resistance index stratified by perinatal conditions: healthy (31 participants, 217 data points), small for gestational age (SGA; 7 participants, 49 data points), large for gestational age (LGA; 3 participants, 21 data points), diabetes (DB; 6 participants, 42 data points), pre-eclampsia (PE; 7 participants, 49 data points), and maternal hypertension (HYN; 10 participants, 70 data points). Each box represents the interquartile range (25^{th} to 75^{th} percentiles), the whiskers denote the 5^{th} to 95^{th} percentile range, and the midline indicates the median. Each participant's data was segmented into 10-min intervals (n = 7 per participant). Each color in the data points corresponds to an individual participant, and the color schemes in **a** and **b** are the same.



Supplementary Fig. 96 | Predictive modeling using logistic regression. Receiver operating characteristic curves from logistic regression analyses for perinatal conditions: a, healthy (31 participants), b, small for gestational age (7 participants), c, large for gestational age (3 participants), d, diabetes (6 participants), e, pre-eclampsia (7 participants), and f, maternal hypertension (10 participants). The plot was constructed using participant-level mean values for pulsatility index, systolic-to-diastolic ratio, resistance index, and fetal heart rate. The dashed diagonal line indicates chance-level performance (AUROC = 0.5). For conditions with limited sample sizes, ROC curves exhibit increased variability, including localized regions below chance-level performance, which is expected for single-feature models constructed from aggregated participant-level data.

Supplementary Video 1 | Real-time tracking. The UPatch can monitor the umbilical cord in the duplex image. The tracking algorithm can register a sample gate on the moving umbilical artery in real time. Then, blood flow spectra can be recorded from the umbilical arteries.

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