DOI 10.5162/Ultrasonic2025/A16-a2

RF Signals from the Superficial Tissues as Stability Index of Ultrasonic Transducer Mounting

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Abstract: Body-mounted ultrasonography stability is critical, as transducer displacement creates RF data artifacts. We quantify mounting stability by analyzing superficial RF signals. Frame-by-frame RF differences were quantified via RMS, defining Instability Waveforms (IW) and depth profiles. Waveforms varied with depth and mounting, exibiting various intensities and regularities, in some cases showing periodicity, potentially from vessel pulsation. A consolidated superficial IW index could be indicative of optimal mounting, improving artifact-free detection of endogenous tissue dynamics.

Keywords: blood vessel, on-body mounted, endogenous pulsation, dynamics imaging.

Introduction

Accurately positioning and sustaining the sensor over a focus area can be difficult and skill-dependent. Such challenges get even more complicated with patient movement or unintentional shifts by the operator, which can compromise data integrity [1]. This instability is critical because the time delays for image reconstruction are directly linked to the distance between the probe and the region of interest. Any movement creates uncertainty [2]. Innovative solutions, such as flexible skin patches, are helping to stabilize wearable ultrasonography devices [1]. However, other broader concerns about maintaining data quality with real-time synchronization and storage plus advanced image processing tools continue to dominate discussions [2]. The areas of reliability, ease of use, and regulatory standards are also areas that remain under-addressed. Given these challenges, the continuous monitoring of mounting stability becomes paramount for ensuring the high quality and reliability of ultrasonography data. In this paper, we propose a new way to measure transducer stability. Our method focuses on analyzing RF signals from the skin's surface and the uppermost tissue layers, an area that's often overlooked. By targeting this specific signal segment, we aim to develop a stability index that reduces artifacts and enhances the accuracy of detecting natural tissue movements.

Background, Motivation and Objective

Body mounted ultrasonic imaging is becoming a new trend [3]. With the body mounted transducer, the in vivo assessment of blood vessels by ultrasonography relies on the stability of acoustic contact. Despite dedicated mounting, the ultrasonic transducer has potential displacements with respect to tissues in vivo, increasing the considerable artifacts in ultrasonic RF data. Special attention to potential instability is considered when dynamic features are characterized in soft tissues. Thus, the monitoring of mounting stability is explored. The objective of the paper is to quantify the stability of transducer mounting by extracting RF data from the skin surface and the superficial tissues. There are published examples of transducer mounting on tissues, but not many provide characteristics of echoscopy signals that are gathered with on-body mounted sensing, particularly in ultrasonography.

Equipment

Signals were obtained with a research-dedicated scanner, Ultrasonix SonixTouch (Analogic Ultrasound, Richmond, BC, Canada), equipped with a 5–14 MHz linear array probe. The application surface area of the linear array transducer was 13 mm x 47 mm. During the ultrasonography on the wrist medial side, imaging was with harmonic 6.6 MHz waves at a frame rate, imaging depth, and focus of 65 fps, 2 cm, and a single focal distance of 0.8 cm, set in the scanner accordingly. During the ultrasonography on the thigh medial side, imaging was with harmonic 5.0 MHz waves at the frame rate, imaging depth, and focus of 66 fps, 3.5 cm, and single focal distance of 0.8 cm, set in the scanner accordingly. A cine-loop stored the data in the form of consecutive frames and RF arrays of digitalized signals for later offline analysis. The holder of the ultrasonic transducer was laboratory-made from hard foam, gluing it with a flexible plastic sheet. The plastic sheet or applicator was 10 cm in length and 16 cm in width.

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In the plastic sheet's central part, the hole was made 20 mm x 55 mm in size. Through this hole, the transducer working surface was coupled to skin. The total weight of the assembled probe was 80 g. On the wrist, the only ultrasonic transducer was stabilized with the laboratory stand, keeping more than 2 mm of gap to the skin. This gap was filled with acoustic coupling gel. The wrist was placed beneath the transducer to ensure blood vessel patterns in the central part of the ultrasonography images. The blood vessels are observed in a transversal direction. Another variant of imaging on the wrist was applying the assembled probe on the wrist to contact the skin with a semirigid surface: 10 cm along the forearm and 6 cm in the transversal direction of the wrist. In the case of imaging in thigh tissues, the whole surface of the applicator was contacting skin. The applicator was applied to the medial side of the thigh to obtain a pattern of femoral blood vessels. The applicator's longer dimension (16 cm) was directed on the thigh in a circumferential direction. Two belts around the thigh were holding the applicator. Belts were possible to bring to adjust compression on tissues.

Methods

We propose analyzing the often neglected part of the RF signal backscattered from the superficial tissue zone as a data source on stability. RF signals from three echolines – central and from both sides of the B-mode – were analyzed from the very first sample. The inter-frame difference was calculated for each echoscopy line Eq. (1).

$$D_L(i) = RF_{L,n}(i) - RF_{L,n+1}(i), \qquad (1)$$

here, RF is an array of radiofrequency data of one ultrasonography frame, L – the index of echoscopy lines, and n – is index of frames in sequence. The time waveforms of interframe difference were evaluated with depth windowing Eq. (2) to estimate depth profiles of activity:

$$Wz_{i,n} = rms(D_{[i-\frac{z}{2}:i+\frac{z}{2},n]}), \qquad (2)$$

here rms(*) – function calculating the root mean square of data; i – the index of locations dividing the first 5 mm depth of imaging into ten zones; z – length of one zone in space, which was 26 RF samples (or 0.5 mm).

Extraction of offset from waveform was made by substracting the mean value Eq. (3):

$$W_{i,n} = Wz_{i,n} - mean(Wz_{i,n}), \qquad (3)$$

here, mean(*) – a function calculating the arithmetic average value of data; Dividing imaging depth into zones enables analysis of tissue dynamics activity at specific depth locations. Estimating the inter-frame differences of RF signals in a sequence of frames provides a dynamic waveform as a function of time. So dynamic waveforms of instability are characterizing tissue activity in any of the locations. Locations in a B-mode image can be arranged with a rectangular grid of any density. In this example the twenty zones are covering the total depth of 5 mm in superficial tissues.

Imaging data

Application sites of the ultrasonic probe with holder were two locations on the researcher's body: the medial side of the wrist and the medial side of the thigh. The first location was preferred because of very superficial blood vessels. The radial artery and two veins were observed in the center of the images. The second location was on massive soft tissues of the thigh, where blood vessels are deep, surrounded by muscles. The great saphenous vein and femoral artery were observed in the image. The sample images from both locations are presented in Fig. 1.

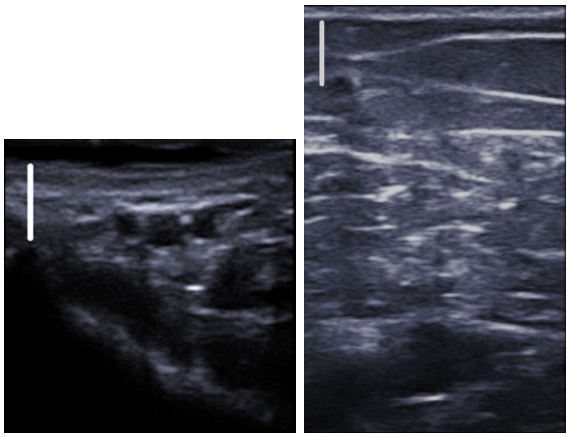

Fig. 1: Tissues and vessels images: a) without direct transducer contact to wrist skin; b) in direct transducer contact to thigh skin. The white line represents the 5 mm scale.

Imaging data for analysis of dynamics in superficial tissues was collected with two fixation options of array transducer. One fixation was on the laboratory stand when the transducer was stabilized in respect to the tabletop. The arm was also stabilized on the tabletop beneath the transducer facing downwards. Acoustic coupling of the medial side of the wrist was with an excessive amount of gel, so the transducer working surface did not touch skin. Transmission of ultrasound waves was established through a gap filled with acoustic coupling gel. The resulting image example is depicted in Fig. 1a.The thickness of the acoustic

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coupling gel was more than 2 mm. The wrist was placed beneath the transducer to ensure blood vessel patterns in the central part of the images. The blood vessels are observed in a transversal view. One, the central vessel appeared to be an artery, as its regular pulsation was observed. With a large applicator on thigh tissues, tightening with belts was interchanged from weak to stronger so that the great saphenous vein initially was observed at 9 mm depth, but after tightening, it was observed at 6 mm depth. The later case of tightening resulted in the image as the example provided in Fig. 1b.

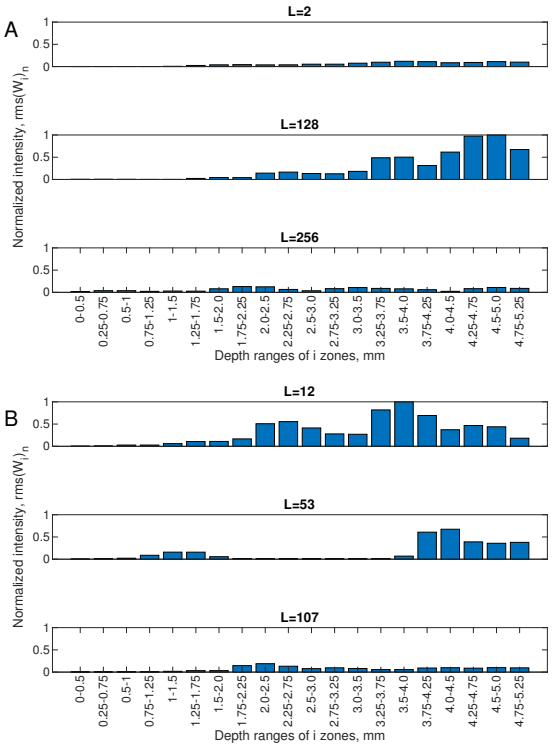

Fig. 2: Intensity of W_i in wrist tissues activity depth profiles at zones of depth: a) coupling with gel-filled gap 2mm; b) no gap coupling.

Results on dynamics of tissues

Common view of pattern in tissues is still grayscale B-mode images as in Fig. 1. We propose diagrams depicting in vivo activity of tissues.

The overview of tissue activity in space of imaging is provided by analyzing $D_L(i)$ at central and at both sides of the B-mode. The activity depth profiles calculated at these three echoscopy lines are presented in Fig. 2. The strongest magnitudes of activity are observed at echoscopy line L=128, z=[4.25-4.75] mm (see Fig. 2a) and at L=12, z=[3.5-4.0] mm (see Fig. 2b). Only the magnitude of activity is depicted, while waveform shapes themselves are addressed in Fig. 3.

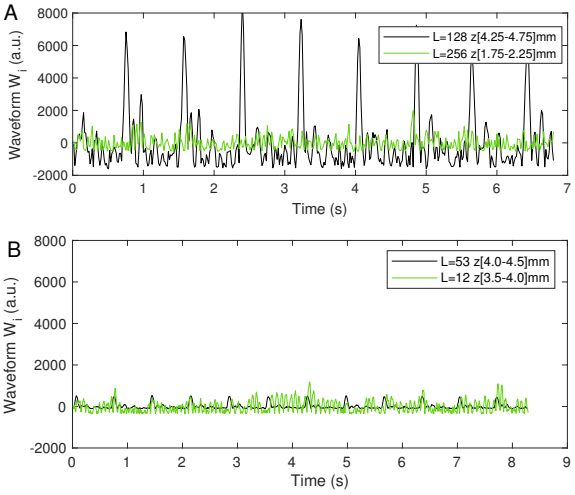

Fig. 3: Instability waveforms evaluated at depth ranges in the wrist: a) with more 2 mm gap at echoscopy line L=128; b) with no gap at L=53.

Comparing tissue activity magnitudes when imaging through a gel-filled gap and when the transducer is in direct coupling to the skin, we observe $D_L(i)$ magnitudes with a ratio around 10 times (5500 vs 520). With a large applicator on thigh tissues, tightening with belts was interchanged from weak to stronger so that the great saphenous vein initially was observed at 9 mm depth, but after tightening, it was observed at 6 mm depth. The intensity-depth profiles in tissues of the thigh are shown in Fig. 4. Identified the most active locations are at L=113, z=[1-1.5] mm (Fig. 4a weak compression with applicator) in case of weakest compression, and L=212, z=[2.25-2.75] mm (Fig. 4b thight compression). Observing the waveforms in Fig. 5 at these locations, we obtain that with weaker applicator compression, we get stronger in vivo activity waveforms with a ratio around 2 times (300 vs 150). The intensity-depth profiles in all cases (Fig. 2 and Fig. 4) accounts for the random activity in tissues, not only physiological pulsing waveforms presented in Fig. 3 and Fig. 5 in black color lines. The random motion artifacts corrupted waveforms are presented in green color lines (see Fig. 3 and Fig. 5), these waveforms are possibly of the magnitude order comparable to physiological waveforms magnitude.

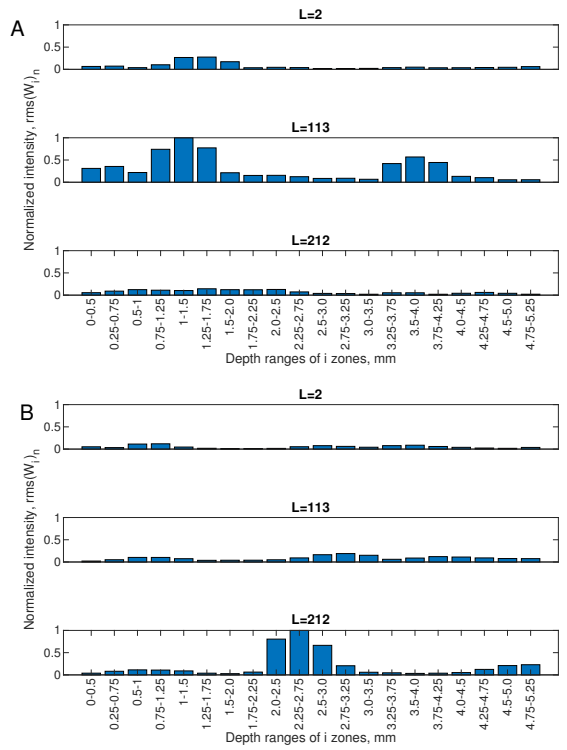

Fig. 4: Intensity of W_i at zones of depth in thigh tissues: a) weak compression (distance to saphenous vein 9 mm); b) stronger compression (distance to saphenous vein 6 mm)

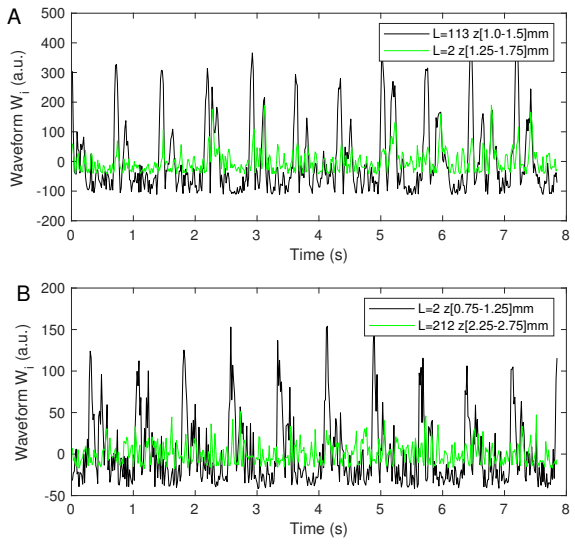

Fig. 5: Instability waveforms evaluated at particular depth ranges in the thigh: a) weak compression (distance to saphenous vein 9 mm); b) stronger compression (distance to saphenous vein 6 mm)

Discussions

Summarizing, instability depth profiles were compared in a few fixing variants of imaging transducers for acoustic coupling to the human body. Instability waveforms appeared to depend on location in tissues and on mounting variants. When the imaging probe is softly mounted physiologically, the arteries induce waveforms of pulsing that are observed. In these cases, high-intensity and highly repeatable in rhythm and shape waveforms are beneficial for the quality of ultrasonography data. But in generalized cases, the instability waveforms $W_{i,n}$ should be analyzed considering their random character. If it is not possible to find a regular rhythm or the inter-waveform correlation is too weak, then it is worth aggregating the calculation of the index of mounting stability. Instability magnitudes $Wz_{i,n}$ at a few echoscopy lines aggregated as a minimal index could be indicative of the maximum mounting stability. But it is important to not compromise the possible damping of highly repeatable waveforms that could be physiological pulsing in tissues induced by arteries.

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