# 2P5-11

# Effect of harmonic imaging on velocity estimation using 2D phase-sensitive motion estimator

Hitoshi Hirano[1‡], Rikuto Suzuki[1], Masaaki Omura [3], Ryo Nagaoka[3], Kozue Saito[2], and Hideyuki Hasegawa[3*]
([1] Graduate School of Science and Engineering, Univ. of Toyama; [2] Department of Neurology, Stroke Center, Nara Medical Univ.; [3] Faculty of Engineering, Univ. of Toyama)

## 1. Introduction

Non-invasive dynamic measurement of the carotid artery wall by ultrasound is used for the diagnosis of atherosclerosis. Although high-frame-rate imaging with unfocused transmit beams is widely used for measurement of cardiovascular dynamics in recent years[1)], high level of clutter and short penetration depth are major drawbacks. Tissue harmonic imaging has been used to reduce clutter in soft tissue measurements such as vessel walls[2)] and a promising method to overcome such problems. In our previous study, we investigated high-frame-rate imaging method using a focused beam for vascular flow imaging[3)].

In this study, we examined high-frame-rate transmit/receive sequences that can be used for harmonic imaging and provide high quality B-mode images and velocity estimation of tissue and blood flow. In addition, we consider the impact of harmonic imaging on velocity estimation.

## 2. Methods

### 2.1 Harmonic imaging

The pulse inversion (PI) method is one of the methods for selectively extracting harmonic components from received echoes. When an ultrasound beam passes through a tissue, the waveform is distorted due to the non-linearity of the tissue, generating harmonic wave signals. The phase of the second harmonic component is changed by $2\pi$ when the phase of the fundamental component is changed by $\pi$. Therefore, if two pulses at a fundamental frequency with a phase difference of $\pi$ are transmitted and the received signals are added, the fundamental component is cancelled and the second harmonic components are enhanced. Harmonic wave signals are generated only in regions where the ultrasonic sound pressure is high. In other words, they are generated only around the central axis of the focused beam, so fewer artefacts due to the influence of surrounding tissue are generated compared to the fundamental component.

‡m23c1037@ems.u-toyama.ac.jp
*hasegawa@eng.u-toyama.ac.jp

### 2.2 Velocity Estimation Method

In this study, the phase-sensitive 2D motion estimator developed by our group was used as a velocity estimate method for the vessel wall[3)].

If we assume that the shape of the radio-frequency (RF) signal from a target remains unchanged from the $n$th frame to the $(n+K)$th frame, the mean squared difference $\alpha$ between the phase $\phi(\omega_x, \omega_z)$ of the cross spectra obtained from echoes in two frames and the phase model is expressed as follows:

$$\alpha = \sum_{\omega_x,\ \omega_z} w(\omega_x,\ \omega_z)\left|\phi(\omega_x,\ \omega_z) + (\omega_x u_x + \omega_z u_z)\right|^2, \quad (1)$$

where $u_x$ and $u_z$ are the lateral and axial displacements, respectively, and $\omega_x$ and $\omega_z$ are the lateral and axial angular frequencies, respectively, and $w(\omega_x, \omega_z)$ is a weight function. The least-squares solution to Eq. (1) is obtained by setting the partial derivatives of $\alpha$ with respect to $u_x$ and $u_z$ to zero to determine $\hat{u}_x$ and $\hat{u}_z$ that minimize the mean squared difference $\alpha$.

### 2.3 Transmit-receive sequence

In our previous study, the frame rate was improved by transmitting two focused beams consecutively, and two receive lines are obtained from one transmit beam using dynamic focusing[3,5)]. In this study, the number of the simultaneously transmitted beams was set to three and four, that was called multi-line transmission (MLT). For 3 and 4 MLT, respectively, the transmit and receive apertures were composed of 36 and 30 elements to avoid interference. Three or four non-steered transmit beams, created in parallel, were focused at a depth of 20 mm. Those focused beams were transmitted three times at the same aperture positions before changing the aperture positions. In the third transmission at each aperture position, the polarity of the signal applied to each transducer element was inverted to perform PI.

The frame rates in 3MLT and 4MLT were 173 Hz and 231 Hz, respectively.

In reception, two receive lines were created based on delay-and-sum (DAS) beamforming with dynamic focusing. At each spatial point along each receive lines, beamformed signals were created from 3 different directions.

## 2.4 Experimental method

**Fig. 1** shows an experimental system using an automatic stage being moved and measured with an ultrasonic probe for velocity estimation. The chicken thigh and skin were placed in front of the probe, the chicken thigh and sound-absorber were placed on the automatic stage and moved at -1.5 mm/s in the depth ($z$) direction and 1.5 mm/s in the lateral ($x$) direction. In a B-mode image, a point of interest was set on the chicken thigh moved by the automatic stage, and velocity estimation was performed using the phase-sensitive 2D motion estimator[3]. The number of simultaneous transmission lines was set at 3 and 4. The summed signal of the first and second transmissions at the same aperture was used for fundamental imaging, and the summed signal of the second and third transmissions was used for harmonic imaging.

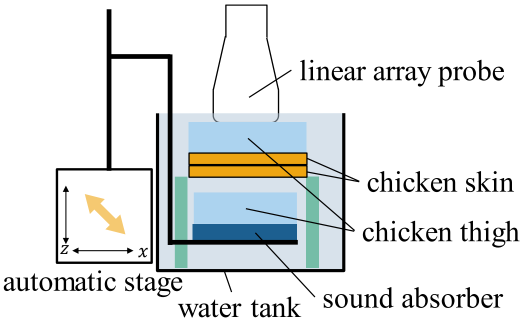


Fig. 1 Illustration of experimental setup.

## 3. Results and discussion

The root mean square error (RMSE) was used as an evaluation index for the velocity estimation results. RMSE in the estimated velocity were evaluated as

$$\mathrm{RMSE} = \sqrt{\frac{\sum(\hat{v}_\beta - v_\beta)^2}{n}}, \qquad (2)$$

where $\hat{v}_\beta$ is the estimated velocity, $v_\beta$ is the true velocity, $n$ is the number of frames, and $\beta$ is $x$ or $z$.

**Table I and II** show the RMSEs of the velocity estimation results. The 4MLT sequence achieves a higher accuracy in the lateral direction than the 3MLT sequence. The higher frame rate was considered to be one of the reasons for the better lateral velocity estimation results in 4MLT than in 3MLT. The velocity estimation results did not change significantly when the PI method was used, while it was expected that the harmonic imaging, which used higher frequencies compared to the fundamental, would improve the accuracy of velocity estimation. We believe this is because the signal-to-noise ratio of the harmonic signal was lower than the fundamental signal, and the bandwidth of the probe was limited.

Table. I RMSE of 3MLT

| | Fundamental [mm/s] | Harmonic [mm/s] |
|---|---|---|
| X | 0.1753 | 0.1847 |
| Z | 0.0568 | 0.0540 |

Table. II RMSE of 4MLT

| | Fundamental [mm/s] | Harmonic [mm/s] |
|---|---|---|
| X | 0.1385 | 0.1655 |
| Z | 0.0561 | 0.0589 |

## 4. Conclusion

In this study, we proposed a high frame rate transmit-receive sequence that can perform both harmonic imaging and tissue and blood flow velocity estimation and investigated the effect of harmonic imaging on tissue velocity estimation. As a result, there was no significant difference between fundamental and harmonic imaging in velocity estimation using the 2D phase-sensitive motion estimator. In the lateral direction, the estimation accuracy was higher for the 4MLT with a higher frame rate.

## References

1) J. Udesen, et al., IEEE Trans. Ultrason. Ferroelectr. Freq. Control, **55**, 1729 (2008).
2) F. Tranquart, et al., Ultrasound Med Biol., **25**, 6, 889 (1999).
3) H. Hasegawa, et al., Jpn. J. Appl. Phys. **61**, SG1026 (2022).
4) H. Hasegawa et al., Appl. Sci. **6**, 195 (2016).
5) L. Tong, et al., 2012 IEEE Int. Ultrason. Symp., 2012, p. 1271.