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High PRF ultrafast sliding compound doppler imaging: fully qualitative and quantitative analysis of blood flow

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Abstract
Ultrafast compound Doppler imaging based on plane-wave excitation (UCDI) can be used to evaluate cardiovascular diseases using high frame rates. In particular, it provides a fully quantifiable flow analysis over a large region of interest with high spatio-temporal resolution. However, the pulse-repetition frequency (PRF) in the UCDI method is limited for high-velocity flow imaging since it has a tradeoff between the number of plane-wave angles (N) and acquisition time. In this paper, we present high PRF ultrafast sliding compound Doppler imaging method (HUSDI) to improve quantitative flow analysis. With the HUSDI method, full scanline images (i.e. each tilted plane wave data) in a Doppler frame buffer are consecutively summed using a sliding window to create high-quality ensemble data so that there is no reduction in frame rate and flow sensitivity. In addition, by updating a new compounding set with a certain time difference (i.e. sliding window step size or L), the HUSDI method allows various Doppler PRFs with the same acquisition data to enable a fully qualitative, retrospective flow assessment. To evaluate the performance of the proposed HUSDI method, simulation, in vitro and in vivo studies were conducted under diverse flow circumstances. In the simulation and in vitro studies, the HUSDI method showed improved hemodynamic representations without reducing either temporal resolution or sensitivity compared to the UCDI method. For the quantitative analysis, the root mean squared velocity error (RMSVE) was measured using 9 angles (−12° to 12°) with L of 1–9, and the results were found to be comparable to those of the UCDI method (L = N = 9), i.e. ≤0.24 cm s⁻¹, for all L values. For the in vivo study, the flow data acquired from a full cardiac cycle of the femoral vessels of a healthy volunteer were analyzed using a PW spectrogram, and arterial and venous flows were successfully assessed with high Doppler PRF (e.g. 5 kHz at L = 4). These results indicate that the proposed HUSDI method can improve flow visualization and quantification with a higher frame rate, PRF and flow sensitivity in cardiovascular imaging.

1. Introduction

Ultrasound Doppler imaging, which provides physiological information (i.e. blood flow direction and velocity), has been widely used to study the cardiovascular system in real time. Over the past several decades, Doppler-based imaging techniques have also been found to be of great value for assessing blood flow through the heart, arteries and veins and may also be used to visualize the motion of solid tissues such as cardiac muscles. Under many clinical conditions, this real-time flow imaging and quantification tool has presented advantages when used in other medical disciplines such as obstetrics and gynecology (Fleischer and Andreotti 2005, Pozniak and Allan 2013).

Among the various ultrasound Doppler imaging techniques, spectral Doppler imaging with continuous wave and pulsed wave (CW and PW) modes (Pozniak and Allan 2013) has been commonly used as a representa-
tive flow quantification tool. The PW Doppler provides quantitative analysis of the flow at a specific site (i.e. sample volume) in the vessel under investigation while the CW Doppler constantly transmits and receives ultrasound waves to assess fast flows with greater sensitivity by sacrificing range information. The spectral Doppler imaging technique also offers quantitative indices such as the spectral broadening index, pulsatility index and resistivity index within a few cardiac cycles (Chavhan et al 2008).

Unlike spectral Doppler imaging, color Doppler imaging can produce a 2D color-encoded map of flow Doppler shifts superimposed onto a B-mode ultrasound image. It can be used to identify blood vessels for examination by using the presence and direction of a flow to highlight local circulation abnormalities found within a region of interest (ROI) (Mitchell 1990). However, color Doppler imaging is inherently limited due to restricted pulse-repetition frequency (PRF) and ROI size, resulting from a finite sound speed. For example, temporal ambiguity can occur when it fails to represent rapid hemodynamic events because the frame rate, which is derived from a desirable Doppler PRF (a few kHz) and ROI size, is low compared to rapidly changing hemodynamic phenomena. In addition, it is often prone to an aliasing artifact when there is a low PRF in the presence of a high-velocity flow. In some cases, sensitivity to low-velocity flows in color Doppler imaging can be mitigated by the short length of the temporal observation window used and by applying a clutter filter with wider transition bands.

Moreover, due to the limited spatial sampling for duplex (B and PW) and triplex (B, Color and PW) modes in commercial ultrasound systems, the performance of each mode generally decreases when the modes are used concurrently, resulting in temporal ambiguity, range ambiguity and aliasing (Maulik 2005). In an effort to overcome inherent tradeoffs between image resolutions and frame rates, a substantial number of studies have been conducted. One approach is to decrease the acquisition time for a frame of color Doppler imaging where acquisition and processing are performed sequentially line by line. For example, a broad focused beam is spatially emitted by interleaving transmit events, and the image is reconstructed by multiple receive beams (Allison 1987). Other approaches involve emitting a few transmit beams or spatially encoded pulses simultaneously and performing parallel receive beamforming or multiline processing (Odershede et al 2008, Cikes et al 2014). In particular, synthetic aperture imaging methods can be applied to flow imaging by producing low-resolution images with a small number of emissions and then combining them coherently to improve image quality (Nikolov and Jensen 2003, Jensen 2012). Recently, plane-wave compounding-based ultrasound imaging with fully parallelized processing architectures known as ultrafast compound imaging has been used in various biomedical applications (e.g. shear wave elastography, natural waves imaging, Doppler flow imaging, and contrast imaging) (Tanter and Fink 2014). Using ultrafast compound imaging allows one to substantially increase the frame rate (to a few kHz) and to create a fully dynamic and focused image for both transmitting and receiving (Montaldo et al 2009). However, ultrafast imaging still presents some limitations compared to general focused beam-based ultrasound imaging techniques. In terms of spatial resolution and sensitivity, the number of compounding angles should increase to several dozen to acquire comparable results to those of a focused beam so that the frame rate may significantly decrease for deep imaging. In addition, ultrafast imaging is susceptible to non-stationary tissue motion (e.g. cardiac imaging) because several tilted full images are coherently combined. To compensate for this constraint, a motion compensation technique based on cross-correlation has been introduced for myocardial imaging (Denarie et al 2013).

Based on the plane-wave angle-compounding method, ultrafast compound Doppler imaging (UCDI) is very useful for cardiovascular imaging (Bercoff et al 2011). In particular, the UCDI method meets the technical needs, such as higher acquisition rates for high-velocity flow imaging, without sacrificing field of view or spatial resolution. In addition, it allows for the visualization of full color Doppler imaging and for full quantitative spectral analysis regardless of the specific region. It has recently been shown to be comparable to high-frame rate flow imaging modalities such as vector Doppler and speckle flow imaging (Hansen et al 2009, Yiu and Yu 2013, Yiu et al 2014), but it can also underestimate high-velocity flows because coherent compounding acts as a low-pass filter (Ekroll et al 2015).

Therefore, the UCDI method achieves significant improvements in terms of temporal resolutions relative to conventional Doppler imaging. However, the UCDI method shows some constraints in terms of acquisition time and sensitivity due to a finite sound speed. In detail, a tradeoff between Doppler PRF and sensitivity depending on the flow velocity concerned still presents challenges because Doppler PRF is determined by the number of compound angles \( N \) and imaging depth \( PRF_{\text{max}} \) used (i.e. \( PRF_{\text{Doppler}} = \frac{PRF_{\text{max}}}{N} \)).

In this paper, we introduce high PRF ultrafast Doppler imaging method based on the plane-wave sliding-angle compounding method (HUSDI). Through the HUSDI method, full scanline images in a Doppler ensemble frame buffer are consecutively compounded using a sliding window; there is no reduction in frame rate, Doppler PRF or sensitivity. By updating a new compounding set with a certain time difference, the HUSDI method enables a variety of Doppler PRF imaging at the same time so that fully qualitative or quantitative analyses can be conducted retrospectively.
2. Methods

2.1. Ultrafast compound Doppler imaging (UCDI)

In the Nyquist–Shannon sampling theorem (Marks 1991), a sufficient sampling rate (i.e. $\geq$ twice the signal bandwidth) is required for perfect signal reconstruction of band-limited signals. Based on this, a fixed Doppler PRF can be determined by the maximum flow-related Doppler shift frequency ($F_s$) in ultrasound Doppler imaging as follows:

$$\text{PRF}_{\text{Doppler}} \geq 2F_s,$$

(1)

Therefore, when applying general color Doppler techniques, when the estimated mean frequency of a widely used narrowband estimator (e.g. autocorrelation-based phase estimation (Kasai et al 1985)) exceeds half of the Doppler PRF, velocity aliasing would occur.

When following the UCDI method, several tilted plane waves are repeatedly transmitted with a certain PRF ($\text{PRF}_{\text{max}}$) based on a maximum imaging depth, and the set of angles ($N$) is sequentially combined to create ensemble data (Bercoff et al 2011). Therefore, the Doppler PRF or flow PRF is determined by the ratio between $\text{PRF}_{\text{max}}$ and $N$ as follows:

$$\text{PRF}_{\text{Doppler}} = \frac{\text{PRF}_{\text{max}}}{N}.$$  

(2)

Based on equation (2), the maximum number of angles is determined by the desired Doppler PRF. In other words, the performance of flow estimations is limited by the acquisition Doppler PRF (Doppler velocity scale). Table 1 shows some examples of detectable Doppler PRF following the number of angles and the imaging depth (or acquisition time) according to the UCDI method. As shown in table 1, the detectable maximum Doppler PRF and flow velocities significantly decrease when the transmitted set of angles increases for high contrast and sensitivity.

In a previous study (Bercoff et al 2011), the UCDI method involving more than 9 angles of tilted plane waves generated similar results to the focused method in terms of flow visualizations. However, in regards to achieving comparable image quality (over 9 angles) in cases of fast flows, the UCDI method suffers from a low Doppler PRF at a given imaging depth. Therefore, for fast flow estimations, one must decrease the number of angles used to obtain a sufficient Doppler PRF, resulting in low image quality.

2.2. Proposed method: high PRF ultrafast sliding compound Doppler imaging (HUSDI)

Under the proposed HUSDI method, a defined angle set of tilted plane waves is transmitted to the medium, which is similar to the conventional UCDI method, but full scanline images (i.e. each angle data) in a frame buffer are coherently summed by a sliding window with a step size (the sliding window size is equal to the number of plane wave angles). In other words, the coherent compounding for high quality images is sequentially conducted at intervals of the sliding window step size whereas the UCDI only conducts the coherent compounding at intervals of the number of angles. In addition, along with the sliding coherent compounding, each plane wave data in a Doppler frame buffer can be used overlapping when the sliding window step size is smaller than the number of transmission angles. Moreover, the angle compounding is continuously updated with each new set of angle compounding data of a certain time difference; the sliding window approach can provide various sampling rates for the same flow signal in Doppler estimates. Therefore, the Doppler PRF of the HUSDI method is determined by sliding window step size $L$, which is greater than zero:

$$\text{PRF}_{\text{Doppler}} = \text{PRF}_{\text{max}}/L, \ (L > 0).$$

(3)
As seen in equation (3), the HUSDI method supports various Doppler PRF sequences regardless of the number of plane wave angles because the angle compounding is only updated as a new compounding set with a factor of \(1/L\) of \(\text{PRF}_{\max}\) using the same number of angles at the same flow rate.

Figure 1 shows the overall Doppler processing sequence using the HUSDI method with different \(L\) values, and it enables varying phase estimation at the same acquisition time. As shown in figure 1, depending on the \(L\) value, the HUSDI provides flow imaging with different PRFs, e.g., high PRF estimations with small \(L\) values and low PRF estimations for larger \(L\) values. In particular, when \(L\) is equal to the number of excited angles \(N\) \((L = N)\), the HUSDI method is the same as the UCDI method.

Under the HUSDI method, similar to UCDI, the mean frequency estimator of the lag-one autocorrelation function (Kasai et al. 1985), which observes the phase-shift between ensemble signals, is used. In the sliding compound technique, the phase change between neighboring sliding compound frames is decreased as the \(L\) value decreases (i.e. the number of overlapping images increases) due to the higher sampling rate of blood flow signals at a shorten acquisition time. Furthermore, the phase variance of the ensemble data is decreased as the \(L\) value decreases since the highly correlated signal, which is reflected from the adjacent flow scatterers, is sampled.

On the other hand, the HUSDI method with high temporal resolution allows a longer ensemble length by using a same temporal observation window. For example, when \(N = 9\) and 90-total firings (i.e. 90-tilted plane wave images) are selected for a frame of color Doppler imaging, the HUSDI method with \(L = 2\) produces approximately 48 ensemble data whereas 10 ensemble data is yielded under the UCDI method. Table 2 shows an example of parameters used for HUSDI imaging with a sound speed of 1540 m \(\text{s}^{-1}\), an imaging depth of 5.0 cm and a frame rate of 100 Hz. As is shown in table 2, for HUSDI, both the Doppler PRF and ensemble length increase when using smaller \(L\) values over the same acquisition time for a frame of color Doppler imaging. Therefore, under the HUSDI method, a longer ensemble length derived from the sliding compound technique can be used to estimate flow velocities; this enables robust Doppler imaging with higher temporal resolution.

Based on this, the HUSDI method of high temporal resolution can be conducted using longer ensemble data for phase estimations to decrease the error estimated from limited temporal averaging over a given acquisition period. The ensemble length of the HUSDI method is increased by a factor of \(L\) at the same acquisition time, and the maximum ensemble length \((E_{\max})\) for a frame of HUSDI imaging can be determined from the following equation:

\[
E_{\max} = \frac{TF + (L - N)}{L},
\]
where TF is the number of total firings (i.e. total acquisition time $\times PRF_{\text{max}}$. In addition, from equations (4) and (6), the frame rate of the HUSDI method is calculated as

$$FrameRate_{\text{HUSDI}} = \frac{PRF_{\text{max}}}{TF} = \frac{PRF_{\text{max}}}{(E_{\text{max}} - 1) \cdot L + N}. \quad (5)$$

In addition, the HUSDI method allows the UCDI method to extend the Nyquist velocity without reducing $N$ so that it is capable of fast-flow imaging with high resolution and contrast. Moreover, the proposed method enables not only full quantification of flow characteristics but also a full qualitative assessment of color Doppler imaging because flow imaging can be analyzed retrospectively at different velocity scales on the same cardiac cycles. For real-time implementation with B-mode imaging in the HUSDI method, the sliding compound to produce different ensemble data for a Doppler frame is only conducted using the same acquisition data of the UCDI method after the B sequence. Therefore, there is no reduction in frame rates when applying the HUSDI method.

3. Experimental methods

3.1. Simulation study
To evaluate the performance of the proposed HUSDI method, a simulation study was conducted using Field II (Jensen 1996). In the simulation, full scanline RF data were generated from 128-active channels using a 128-element linear array transducer (0.3 mm pitch). The center and sampling frequencies were 5 and 20 MHz, respectively. For plane wave angle compounding, 9-tilted plane waves ($-12^\circ$, $-9^\circ$, $-6^\circ$, $-3^\circ$, $0^\circ$, $3^\circ$, $6^\circ$, $9^\circ$, $12^\circ$) were utilized. The imaging depth was 30 mm according to the $PRF_{\text{max}}$ of 25 kHz and the number of total firings was 180 (i.e. 7.2 ms acquisition time). To perform blood flow imaging, plug and laminar flows using the Hagen-Poiseuille profile (Stuart 1960) were produced (10 scatterers per a resolution cell) within the 5 mm radius of the vessel, and the maximum velocity in the center of the vessel was set to 10, 30 and 50 cm s$^{-1}$ (Holdsworth et al 1999). In addition, the beam-to-flow angle was set to 45$^\circ$, and no clutter filter was applied.

3.2. In vitro phantom study
The performance of the proposed HUSDI method was evaluated under realistic complex flow dynamics of stenotic conditions by using a carotid bifurcation phantom based on the previously proposed rapid prototyping (RP) framework (Lai et al 2013, Yiu and Yu 2013, Leow et al 2015). To fabricate a stenosed bifurcation vessel tube, a stereolithography input file of the vessel-mimicking model was first created using computer-aided design software (SolidWorks; Dassault Systemes, Waltham, MA, USA). The stenosis geometry, including lumen diameters of the common carotid artery (CCA), internal carotid artery (ICA) and external carotid artery (ECA), and a blockage of stenosis are described in a previous study (Lai et al 2013). A non-rigid vessel tube for this model was then produced via a RP machine (Objet Connex500; Objet Geometries, Rehovot, Israel) using a compliant photopolymer material (FullCure 930; Objet Geometries, Rehovot, Israel) and a gel-like supporting photosensitive resin (FullCure 705; Objet Geometries, Rehovot, Israel), as is shown in figure 2(a). In particular, the mechanical properties of the compliant photopolymer material (i.e. vessel tube material) were found to be similar to healthy human arteries with an elastic modulus range of 379–404 kPa (Lai et al 2013).

To construct a full-scale replica of the carotid bifurcation phantom, the resulting compliant vessel tube was aligned with the vertical plane and was mounted on an acrylic box with hose barbs at the ends and an acoustic absorber positioned at the base. Tissue-mimicking material for casting the vessel tube in a box was created based on an agar mixture and was composed of 85.1% distilled water; 3.0% agar (A1296; Sigma-Aldrich, St. Louis, MO, USA); 11.1% glycerol (G7757; Sigma-Aldrich); 0.5% silicon dioxide (S5631; Sigma-Aldrich); and 0.3% potassium sorbate preservative (85520; Sigma-Aldrich), reflecting that used in a previous study (Lai et al

<table>
<thead>
<tr>
<th>PRF$_{\text{max}}$ (kHz)</th>
<th>$L$</th>
<th>PRF$_{\text{Doppler}}$ (kHz)</th>
<th>Ensemble length</th>
<th>Acquisition time (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>15.4</td>
<td>146</td>
<td></td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>7.7</td>
<td>73</td>
<td></td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>5.1</td>
<td>49</td>
<td></td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>3.9</td>
<td>37</td>
<td></td>
<td></td>
</tr>
<tr>
<td>15.4</td>
<td>5</td>
<td>31</td>
<td>10</td>
<td></td>
</tr>
<tr>
<td>6</td>
<td>2.6</td>
<td>25</td>
<td></td>
<td></td>
</tr>
<tr>
<td>7</td>
<td>2.2</td>
<td>21</td>
<td></td>
<td></td>
</tr>
<tr>
<td>8</td>
<td>1.9</td>
<td>19</td>
<td></td>
<td></td>
</tr>
<tr>
<td>9 (UCDI)</td>
<td>1.7</td>
<td>17</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 2. An example of parameters of the proposed HUSDI method with 9 angles.
Additionally, the agar concentration of 3.0% includes the Young’s modulus of about 170 kPa (Manickam et al. 2014). To supply a constant or pulsatile flow to the bifurcation phantom, a programmable gear pump system (CompuFlow 1000MR; Shelley Medical Imaging, London, ON, Canada) with a volume range of 0.01–35 ml s\(^{-1}\), flow waveforms of 1.0–12 Hz, and blood-mimicking fluid (Model 046; CIRS, Norfolk, VA, USA) was used.

Figure 2(b) shows the experimental setup used for the in vitro study.

In the in vitro phantom study, experiments were performed using the anthropomorphic bifurcation phantom with lumen diameters of 6.0, 4.2 and 3.5 mm for the CCA, ICA and ECA, respectively, under an input flow rate of 4 ml s\(^{-1}\). The beam-to-flow angle for deriving flow velocity was set to 75\(^\circ\). Transmit conditions in terms of resolution and contrast for the plane wave compounding method are summarized in table 3, and the acquisition parameters for flow measurements (\(N = 1, 3, 5\) and 9) are summarized in table 4. The ensemble length for each compound case of the in vitro experiments was set to the maximal value corresponding to equation (4) at the 10 ms acquisition time. For the wall filter to suppress unwanted tissue signals, a 5th-order projection-initialized IIR filter (Chornoboy 1992) was applied, and the stop-band frequency was set to 100 Hz in accordance with the \(L\) value because the wall signal is shifted by the flow sampling rate (i.e. Doppler PRF) in the frequency domain. In addition, the HUSDI method allows the wall signal to apply different wall filters with narrower transition bands because a longer ensemble length based on \(L\) can be used. Moreover, the flow images show the results without any type of spatial or temporal filtering. To qualitatively or quantitatively assess the performance of the HUSDI method based on various angle sets, HUSDI images with \(N = 1, 3, 5\) and 9 were processed using the same parameters as those summarized in tables 3 and 4.

### 3.3. In vivo study

To assess the performance of the HUSDI method under fast-flow circumstances, broad-view channel data for a 2 s time period were obtained from a healthy volunteer’s common femoral vessel area at flow velocities of 50–100 cm s\(^{-1}\) under the Institutional Review Board (IRB) approval. For an ultrafast transmission strategy, 9-plane waves with alternate polarity sequencing (\(-12^\circ, 12^\circ, -9^\circ, 9^\circ, -6^\circ, 6^\circ, -3^\circ, 3^\circ, 0^\circ\)) were transmitted to avoid lateral shifts of the moving targets (Denarie et al. 2013) and PRF\(_{\text{max}}\) was set to 20 kHz. The acquisition time between angles or frames was set to 1/PRF\(_{\text{max}}\) for the sliding window compounding approach. Therefore, the detectable Doppler PRF range from the sliding window step size \(L\) under the HUSDI method can range from 2.2 to 20 kHz, while the UCDI method can only estimate a mean Doppler frequency of 2.2 kHz (\(L = N = 9\)) for color.
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and spectral Doppler imaging. In addition, the ensemble length was used as the maximum value corresponding
to a 18 ms acquisition time for a frame of flow images. For the PW spectrograms, the degree of overlap was 1/8
and a 64-point hamming window was utilized. The Doppler angle was manually optimized (it was calculated
assuming that the blood flow was horizontal to the vessel wall). Consequently, the HUSDI method is useful to
qualitatively and quantitatively analyze flow dynamics for the same acquisition data over the same cardiac cycle
simultaneously.

3.4. Data acquisition

All data acquisition for the performance assessment was conducted using a channel domain imaging research
platform (Vantage128, Verasonics Inc., Kirkland, WA, USA). The system was connected with L7-4 (Philips
Healthcare, Andover, MA, USA) linear array transducer (0.298 mm pitch) of 128 elements, which is generally
utilized for vascular imaging. Full scanline channel data for the plane-wave beamforming strategy (Montaldo
et al. 2009) were obtained using unfocused transmit pulses with a center frequency of 5.2 MHz (a wavelength
of 0.296 mm) and a duration of four cycles. The sampling rate was set to 20.8 MHz by 14 bit A/D converters in
the system. Channel data were transferred to an external PC, and beamforming and flow signal processing were
conducted through a custom-built Matlab program (R2016a; Mathworks, Natick, MA, USA). For transmit and
receive events, a programmable user-defined transmit/receive sequence was used according to the experimental
conditions. The image resolution parameters used for plane-wave coherent compounding are described in
table 3.

3.5. Evaluation metrics

For the quantitative performance evaluation, the signal-to-clutter ratio (SCR) and root mean squared velocity
error (RMSVE) were measured. The SCR is defined by

$$\text{SCR}_{\text{dB}} = 10 \log_{10} \frac{\text{Power}_{\text{flow}}}{\text{Power}_{\text{clutter}}},$$

(6)

where Power_{flow} and Power_{clutter} are the average power of flow signals within the flow vessel and the average power
of clutter signals, which is estimated from tissue region.

In addition, the RMSVE can be defined according to the theoretical Poiseuille law, which is derived from
Navier–Stokes equations. The Hagen–Poiseuille profile provides the full solution for laminar and parabolic flows
in a cylindrical tube as well as the pressure drop for an incompressible, Newtonian fluid in a stationary flow (non-
pulsatile) (Stuart 1960). The theoretical Poiseuille flow velocity is measured as:

$$v_p = v_{\text{max}} \left[1 - \left(\frac{r_p}{R}\right)^2\right],$$

(7)

Table 3. Experimental parameters of the plane-wave coherent compounding method.

<table>
<thead>
<tr>
<th>Parameters</th>
<th>N = 1</th>
<th>N = 3</th>
<th>N = 5</th>
<th>N = 9</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\alpha_{\text{width}}$</td>
<td>-</td>
<td>0.42</td>
<td>0.42</td>
<td>0.42</td>
</tr>
<tr>
<td>F_number</td>
<td>2.4</td>
<td>2.4</td>
<td>2.4</td>
<td></td>
</tr>
<tr>
<td>ML_{width} (mm)</td>
<td>0.49</td>
<td>0.39</td>
<td>0.66</td>
<td></td>
</tr>
<tr>
<td>GL_{location} (mm)</td>
<td>1.48</td>
<td>2.96</td>
<td>5.92</td>
<td></td>
</tr>
<tr>
<td>Scanline</td>
<td>128</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Apodization window</td>
<td>Kaiser ($\alpha = 1$)</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

$\alpha_{\text{width}} = \alpha_{\text{max}} - \alpha_{\text{min}}$ ($\alpha = \sin \theta$, $\theta_{\text{steering angle}} = -12^\circ$ to $12^\circ$). ML and GL are the—6 dB main lobe and grating lobe of the pressure field, respectively.

Table 4. Acquisition parameters of the HUSDI method for the in vitro study.

<table>
<thead>
<tr>
<th>PRF_{max} (kHz)</th>
<th>Sliding window step size (L)</th>
<th>N</th>
<th>PRF_{Doppler} (kHz)</th>
<th>Total number of firings</th>
<th>Acquisition time (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>No compound 1</td>
<td>6</td>
<td>3</td>
<td>1.5</td>
<td>0.9</td>
</tr>
<tr>
<td>6</td>
<td>Compound 3</td>
<td>6</td>
<td>3</td>
<td>1.5</td>
<td>1.2</td>
</tr>
<tr>
<td>10</td>
<td>Compound 5</td>
<td>10</td>
<td>5</td>
<td>3.3</td>
<td>2.5</td>
</tr>
<tr>
<td>18</td>
<td>Compound 9</td>
<td>18</td>
<td>9</td>
<td>4.5</td>
<td>3.6</td>
</tr>
</tbody>
</table>

a Denotes the UCDI method.
where \( v_{\text{max}} = \frac{2Q}{\pi R^2} \) is the maximum velocity at the center of the tube, \( r_p \) is the radial location within the tube, \( Q \) is the flow rate and \( R \) is the tube radius. Therefore, the RMSVE is calculated from the experimental flow velocity \( (v_k) \) and theoretical Poiseuille flow velocity from equation (7) as follows:

\[
\text{RMSVE}_{\text{cm s}^{-1}} = \sqrt{\frac{\sum_{k=1}^{n} (\hat{v}_p - v_k)^2}{n}},
\]

where \( n \) is the number of samples.

4. Results

4.1. Simulation study

Figure 3 shows the phase-shift estimation using the HUSDI method \( (N = 9) \) according to the flow velocity 10, 30 and 50 cm s\(^{-1}\) in the plug flow simulation. (a)–(c) The simulation results using the same ensemble length \( (E = 20) \) with respect to \( L \). (d)–(f) The simulation results using the maximum ensemble length \( (E_{\text{max}}) \) with respect to \( L \). The red solid line indicates the true value of the flow velocity.

Figure 4 shows color Doppler images derived from the HUSDI laminar flow simulation \( (v_{\text{max}} = 50 \text{ cm s}^{-1}) \). The same parameters used for the plug flow simulation shown in figure 3 were used to reconstruct color Doppler images. The maximum imaging depth was 30 mm according to the maximum PRF (i.e. 25 kHz) and a sound
speed of 1540 m s\(^{-1}\), and no spatial or temporal filtering was applied. As shown in figures 4(a) and (b), the HUSDI method with higher PRFs (i.e. \(L = 4\) and 5) clearly display laminar flow patterns in the color Doppler images without aliasing. However, as the \(L\) value increases, the PRF in the HUSDI decreases, and an aliasing artifact occurs as shown in figures 4(c) and (d), which are similar to figures 3(c) and (f).

To conduct a quantitative assessment of flow sensitivity with respect to the \(L\) value, the flow power was measured in a plug flow simulation depicted in figure 3 for flow velocities of 10, 30 and 50 cm s\(^{-1}\). The HUSDI method supports approximately 27.0 dB in terms of the SCR (the power of clutter was measured using noise region) and no differences in \(L\) values were found. As was demonstrated in the simulation study, the proposed HUSDI method can dynamically depict fast and slow flows by adjusting the \(L\) value (i.e. different Doppler PRFs) without sacrificing frame rate and sensitivity.

### 4.2. In vitro phantom study

Figure 5 shows the HUSDI imaging results according to the \(L\) value with the maximum ensemble length for a 10 ms acquisition time for the CCA region. As shown in figure 5, the HUSDI method allows the various Doppler

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**Table 5.** The phase-shift estimation results using the 10 images of the HUSDI method (\(N = 9, E = 20\)) in the plug flow simulation study (mean \(\pm\) standard deviation).

<table>
<thead>
<tr>
<th>Sliding window step size ((L))</th>
<th>Plug flow velocity (cm s(^{-1}))</th>
<th>Doppler estimate (phase/2(\pi)) (True value)</th>
</tr>
</thead>
<tbody>
<tr>
<td>10</td>
<td>10</td>
<td>0.02 (\pm) 0.02 (0.02) 0.05 (\pm) 0.01 (0.05) 0.08 (\pm) 0.03 (0.09)</td>
</tr>
<tr>
<td>30</td>
<td>30</td>
<td>0.03 (\pm) 0.02 (0.03) 0.10 (\pm) 0.02 (0.11) 0.16 (\pm) 0.05 (0.18)</td>
</tr>
<tr>
<td>50</td>
<td>50</td>
<td>0.05 (\pm) 0.02 (0.05) 0.15 (\pm) 0.02 (0.16) 0.24 (\pm) 0.06 (0.26)</td>
</tr>
<tr>
<td>10</td>
<td>10</td>
<td>0.07 (\pm) 0.02 (0.07) 0.20 (\pm) 0.03 (0.21) 0.33 (\pm) 0.07 (0.35)</td>
</tr>
<tr>
<td>30</td>
<td>30</td>
<td>0.09 (\pm) 0.02 (0.09) 0.25 (\pm) 0.03 (0.26) 0.40 (\pm) 0.07 (0.44)</td>
</tr>
<tr>
<td>50</td>
<td>50</td>
<td>0.11 (\pm) 0.02 (0.11) 0.30 (\pm) 0.05 (0.32) —0.41 (\pm) 0.08 (−0.47)</td>
</tr>
<tr>
<td>10</td>
<td>10</td>
<td>0.12 (\pm) 0.04 (0.12) 0.36 (\pm) 0.06 (0.37) —0.42 (\pm) 0.08 (−0.38)</td>
</tr>
<tr>
<td>30</td>
<td>30</td>
<td>0.14 (\pm) 0.04 (0.14) 0.41 (\pm) 0.06 (0.42) —0.35 (\pm) 0.09 (−0.29)</td>
</tr>
<tr>
<td>50</td>
<td>50</td>
<td>0.15 (\pm) 0.04 (0.16) 0.46 (\pm) 0.06 (0.48) —0.26 (\pm) 0.01 (−0.21)</td>
</tr>
</tbody>
</table>

**Figure 4.** Color Doppler imaging results derived from the HUSDI method (\(N = 9\)) with different \(L\) values from the laminar flow simulation (50 cm s\(^{-1}\)). (a)–(c) show the results of the diverse Doppler scale and (d) shows Doppler aliasing artifacts with relatively low Doppler PRF values derived from the UCDI method.
PRF images to correspond to different $L$ values and to an increasing number of angles ($N$). In particular, the method can be used to produce high-PRF flow images without reducing the number of angles (i.e. the $L = 2$ image in figure 5(d)), unlike the conventional UCDI method (i.e. the $L = 9$ image in figure 5(d)). The SCR and RMSVE values of the HUSDI images with $N = 1, 3, 5$ and 9 were also measured, the ROI was set to the center of the lumen, and the size was set to $16 \times 16$ pixels ($0.6 \times 0.6$ mm). Figures 6(a) and (b) illustrate the measured SCR and RMSVE results of the 10 images of each angle case. The ground truth of RMSVE was calculated using the theoretical Poiseuille law of equation (7). The measured SCR values with different $L$ values of 1–9 (figure 6(a)) show comparable results to those of the conventional UCDI method (i.e. $L = N$) for all angle sets, and particularly for $N = 9$ (table 6). In other words, the HUSDI method can be used to support a faster Doppler PRF without reducing the number of angles ($N$) relative to the UCDI method. At the same time, the RMSVE values shown in figure 6(b) exhibit comparable results, and particularly for $N = 9$ (table 6). In addition, the RMSVE was mostly decreased for all $L$ values when the number of transmission angles increases as shown in figure 6(b).

To evaluate the complex flow circumstances with disturbed flow pattern generally found in a stenosed bifurcation with plaque ulcerations, the artery branch of a diseased flow phantom was assessed. To compare the performance of the conventional UCDI and proposed HUSDI in regards to fast-flow imaging, the input flow rate was increased to 10 ml s$^{-1}$. The parameters for data acquisition used were the same as those shown in tables 3 and 4. Figure 7 shows the results of the UCDI and HUSDI methods for $N = 9$. A single plane-wave Doppler imaging method ($N = 1$) was also added to compare the image quality and flow sensitivity with the UCDI and HUSDI methods. As shown in figure 7(a), the single plane-wave Doppler imaging exhibits fast-flow imaging at a good temporal resolution (PRF$_{Doppler} = 6$ kHz), but the image quality degraded significantly in terms of contrast and sensitivity compared to $N = 9$ because only single plane-wave excitation was used for flow estimations. While a
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UCDI image with \( N = L = 9 \) (PRF\textsubscript{Doppler} = 2 kHz) can enhance the sensitivity of flow visualizations as is shown in figure 7(b), it yields a pronounced aliasing artifact at the entire flows due to the limited Doppler PRF. Due to the inherent trade-off between SNR and acquisition time, it is difficult to obtain a high-quality image under fast-flow conditions using the UCDI method. Figures 7(c) and (d) show color Doppler images derived from the HUSDI method with \( L = 3 \) (i.e. PRF\textsubscript{Doppler} = 6 kHz) and \( L = 2 \) (i.e. 9 kHz), respectively, using the same set of data shown in figure 7(b). As shown in figures 7(c) and (d), the HUSDI approach generates high-quality flow images without a loss in the number of angles and thus avoids the aliasing artifact, and it also improves the visualization of flow dynamics considerably. For example, flow jets (under the yellow arrow) of the stenosis site at the ICA inlet and recirculation flows (upper yellow arrow) from post-stenotic flows at the carotid bulb are clearly shown in figures 7(c) and (d). In addition, a reverse flow at the ECA inlet formed as a result of beam flow angle dependency, and the outflow of the ECA shows strong sensitivity, excluding the shadowing region of the ECA.

<table>
<thead>
<tr>
<th>Sliding window step size (L)</th>
<th>Evaluation metrics</th>
</tr>
</thead>
<tbody>
<tr>
<td>SCR (dB)</td>
<td>RMSVE (cm s(^{-1}))</td>
</tr>
<tr>
<td>1</td>
<td>22.4 ± 0.7</td>
</tr>
<tr>
<td>2</td>
<td>23.7 ± 0.2</td>
</tr>
<tr>
<td>3</td>
<td>24.2 ± 0.1</td>
</tr>
<tr>
<td>4</td>
<td>24.0 ± 0.1</td>
</tr>
<tr>
<td>5</td>
<td>24.0 ± 0.2</td>
</tr>
<tr>
<td>6</td>
<td>24.1 ± 0.2</td>
</tr>
<tr>
<td>7</td>
<td>24.5 ± 0.1</td>
</tr>
<tr>
<td>8</td>
<td>24.9 ± 0.1</td>
</tr>
<tr>
<td>9</td>
<td>24.4 ± 0.2</td>
</tr>
</tbody>
</table>

Table 6. The measured SCR and RMSVE values with the 10 images of the HUSDI method (\( N = 9 \)) at the CCA region in the carotid bifurcation phantom (mean ± standard deviation).
inlet (white arrow) resulting from severe tissue attenuation (Undesirable bubbles were added at the branch site during tissue phantom fabrication). In addition, the HUSDI method outperforms the UCDI method in terms of qualitative analysis, as it generates various Doppler PRF flow images relative to the desired Doppler velocity scale as is shown in figure 7(d).

4.3. In vivo study

Figure 8 illustrates the HUSDI results with the ultrafast data, which is obtained from the femoral vessel area (SFA: superficial femoral artery, CFV: common femoral vein and DFA: deep femoral artery). For an ultrafast data acquisition, 9-plane waves were utilized and \( PRF_{\text{max}} \) was set to 20 kHz. Multi-sample volume spectrograms corresponding to each vessel were also represented \((N = L = 9)\) and they exhibit different flow velocity and direction in a full cardiac cycle as shown in figure 8(a). Figure 8(b) shows interesting phases of a full cardiac cycle \((A, B \text{ and } C)\) when different \( L \) was retrospectively applied in the HUSDI method. Unlike the UCDI method, the HUSDI method enables the diverse flow images in each \( A, B \text{ and } C \) phase at the short acquisition time. At the \( A \) phase in the figure 8(a), the antegrade flow corresponding to the second component of the spectral waveform during the mid-systole period occurred in SFA and DFA as the CFV approached peak velocity flows. In particular, the venous flow at the bifurcation site in the CFV shows the aliasing velocity as the sliding \( L \) value increases very high \((L = 14)\); it can also improve the visualization of flow dynamics.

![Diagram showing HUSDI results](image)

**Figure 8.** (a) The HUSDI sequence using the ultrafast data obtained from a femoral artery and vein segment (SFA: superficial femoral artery, CFV: common femoral artery and DFA: deep femoral artery) and the simultaneous PW spectrograms from the sample volumes in a proximal lumen of three vessels. (b) The HUSDI images derived from different \( L \) values for the same acquisition time (i.e. 18 ms) at the specific phases of a full cardiac cycle \((A, B \text{ and } C)\). This method can enable diverse complex flow visualization according to the \( L \) without causing a loss of sensitivity or acquisition time. The * denotes the UCDI method \((L = N)\).
The B phase shows the first-time component corresponding to rapid antegrade flows measured during nearby peak systole in the SFA and DFA relative to the CFV. The arterial flow in the DFA accelerated at a faster rate than it did in the SFA, and the flow image of $L = 9$ (UCDI method) in the B phase illustrates severe velocity aliasing in the DFA due to limited Doppler PRF ($20/9 = 2.2$ kHz). However, by controlling the $L$ value to be smaller on the same acquisition data, a color Doppler image with a higher PRF can obtain the optimal velocity estimation without severe aliasing artifacts as shown in flow images from $L = 4$ to 6. The C phase images show that three vessels followed a reversal flow at the same time and various scaled reverse flow images were represented by using the ultrafast HUSDI method. Also, such instantaneous flow emergence can be only observed with ultrafast Doppler imaging because standard color Doppler imaging is limited to frame rates of up to 20–30 Hz.

Figure 9 demonstrates the example of the retrospective flow analysis using the HUSDI method. At the top in the figure 9, consecutive pictures shows several interesting frames for a single cardiac cycle and it also provides some momentary complex flows resulting from the rapid inversions of a cardiac cycle; this was only found when applying the ultrafast Doppler method. Also, these instantaneous complex flow and transient flow events require very short acquisition periods and high sensitivity. Figures 9(a) and (b) shows the procedure and results of arterial blood flow (SFA and DFA) and venous flow (CFV) analysis. For retrospective flow analysis in the HUSDI method, interesting frames are manually selected from the cineloop and qualitative analysis of complex flow dynamics is performed by controlling the $L$ value (from low PRF to high PRF). Then, multiple interesting regions in the flow images are simultaneously selected and quantified flow velocities at various velocity scales. Note that

Figure 9. Representative HUSDI frames of the full cardiac cycle of a femoral artery and vein, and the example of the retrospective flow analysis using the HUSDI method. This approach provides different velocity-scaled Doppler images and full quantifications without a loss of sensitivity or temporal resolution with transition time through sliding window lag (i.e. sliding window step size) control.
the HUSDI method enables simultaneous qualitative and quantitative flow analysis on the same acquisition data without new pulsed-wave excitation efforts. Figure 9(a) shows arterial blood flow analysis during nearby peak systole in the SFA and DFA. Interesting two sample volumes (1 × 1 mm) at the center of the lumen in each artery were selected and PW spectrograms during a second were depicted for velocity quantification. As shown in figure 9(a), the arterial segment (SFA and DFA) exhibits a triphasic flow pattern due to the presence of high levels of peripheral resistance under healthy conditions. As changing user selective L value, improved PW spectra without Doppler aliasing (\( N = L = 9 \)) for a complete cardiac cycle were successfully produced due to higher Doppler PRF despite the fact that a baseline shift was not used. For the venous blood flow, two sample volumes at the common (1) and bifurcation site (2) in the CFV were selected as illustrated in figure 9(b). Unlike the arterial blood flow, monophasic waveform was mainly shown according to the venous characteristics, which is reliable indicators of proximal venous obstruction in the common femoral vein. In addition, as shown in quantification results of PW spectra, the disturbed flow at the bifurcation site (2) was higher than the flow of the common site (1). In addition, instantaneous forward flow from rapid inversion of a cardiac cycle in both spectra was clearly observed due to high temporal resolutions achieved.

Consequently, the HUSDI method, which supports a high spatio-temporal resolution, can be used to provide more reliable flow visualization and full quantifications without a loss of temporal information (i.e. temporal ambiguity) with transition time. In addition, unlike the conventional UCDI method, the HUSDI method enables one to conduct qualitative flow analysis using a sliding compound strategy based on different time values (L). Therefore, it can be used to achieve fine-to-coarse scale flow imaging without compromising sensitivity or acquisition time for the same data and time retrospectively.

5. Discussion

UCDI has an advantage of high quality flow estimations with significantly short acquisition time as well as full ROI quantification. However, it comes with a tradeoff between acquisition time and sensitivity as Doppler PRF is determined from the number of plane wave transmissions with different tilted angles. Here, we have proposed an ultrafast sliding compound Doppler imaging method that can be used to provide high Doppler PRF and high sensitivity without reducing the frame rate. In using this method, which is similar to the conventional ultrafast Doppler approach, full scanline RF data in a frame buffer are coherently compounded, but angle compounding is performed by a sliding window of a different step size. Therefore, the Doppler PRF is determined by the sliding window step size rather than by the number of plane wave angles involved. To derive various Doppler PRFs relative to flow velocities using this sliding approach, two major technical elements must be applied: (1) the acquisition time between plane wave angles and frames for ensemble data should be same and (2) the ensemble length should be increased to compensate for the error in the estimated velocity. In other words, all flow information for the acquisition time can be included when the maximum ensemble length is used for velocity estimation. This simple approach can thus be used to carry out diverse high sensitivity flow imaging from fast-to slow-flow estimations; it enables simultaneously qualitative and quantitative flow analysis in a retrospective manner.

As a clinical use, the proposed method could be used to facilitate the adequate detection of subtle changes in morphological flow dynamics as an indicator of preclinical atherosclerosis or plaque severity. In particular, flow velocity or pressure around atherosclerotic plaques are significantly increased by a narrowed lumen, and so the measurable velocity also should be high when identifying complex flows. In addition, the method can reduce the number of operational considerations required (e.g. Doppler PRF adjustments), as it can derive optimal flow images from a retrospective flow analysis without rescanning. On the other hand, the technique can also be used in various cardiovascular assessments requiring short acquisition times (e.g. arterial motion estimation (Salles et al 2015), vascular strain compounding (Hansen et al 2014) and pulse wave velocity (PWV) measurements (Couade et al 2011) for vessels, and slow-flow imaging applications (Mace et al 2011, Couture et al 2012)).

The ultrafast sliding compound Doppler method can be used for high-frame rate Doppler imaging, and it can be used as an effective cardiovascular evaluation tool. However, the approach is also inherently limited by its use of coherent compounding techniques. Retrospective transmit focusing methods such as synthetic transmit aperture imaging (Gammelmark and Jensen 2014) or coherent plane wave compounding (Denarie et al 2013) are susceptible to tissue motion, as both methods perform a coherent summation of received signals to achieve increased SNR and a high-spatial resolution point spread function (PSF). Similarly, for blood motion in coherent compound Doppler imaging, the compounded flow signal at the Doppler PRF may produce negatively biased velocities depending on transmit and receive directions, and the SNR is also degraded as a result of out-of-phase summation (Ekroll et al 2015). Therefore, motion compensation methods and different transmit or receive sequences can reduce the effects of blood motion, thus mitigating negative biases of flow estimates and SNR.

In the same manner, tissue motion in UCDI inhibits well-defined PSF since the scatterers move between each plane wave transmission. In addition, for sliding coherent compounding, degraded PSFs produce different
sidelobe levels for each compounding frame. In the slow-time processing for Doppler imaging, these sidelobes exhibit a periodicity in accordance with the number of plane-wave angles in several ensemble frames and clutter artifacts may occur. To overcome this issue, a clutter artifact rejection algorithm based on spectral analysis (e.g. singular value decomposition) is currently under research and development.

Based on the sliding compound Doppler imaging method, some points remain to be clarified such as how to manage the frame buffer for qualitative or quantitative analysis using a real-time ultrasound system or how helpful the proposed method is for real clinical evaluations. In future work, for thorough technical verification, the proposed method will be evaluated with CFD (computational fluid dynamics) simulation or PIV (particle image velocimetry) using microbubbles. In addition, an efficient method for the simultaneous qualitative and quantitative analysis of the cardiovascular system will be addressed and additional in vivo experiments for various diseased vessels (e.g. atherosclerosis) will be conducted. Moreover, beyond effects of ultrafast Doppler imaging, the effects of arterial wall motion in an ultrafast sliding sequence will be analyzed and applied in very high frame rate applications (e.g. PWV imaging (Couade et al 2011) or electromechanical wave detection (Prowost et al 2013)).

6. Conclusion

High-frame rate Doppler flow imaging is very useful for evaluating cardiovascular diagnoses and flow characteristics. In this paper, we present HUSDI method that could improve complex flow analysis without a loss of sensitivity or acquisition time. Unlike conventional UCDI, the angle compounding in the HUSDI method is consecutively updated with a certain time difference to allow for various Doppler frequencies without an inherent trade-off between the number of angles and frame rates. The proposed method thus supports advanced flow image analysis as well as full ROI quantification by generating various scaled Doppler images from the same acquisition data and at the same time. Based on the various flow conditions (e.g. diseased vessels and diverse velocity flows) of our simulation, in vitro and in vivo studies, we have shown that the HUSDI method can be used to depict complex flow dynamics from a laminar flow profile at a high spatio-temporal resolution. Consequently, the HUSDI method is expected to provide more accurate and reliable hemodynamic assessments from qualitative or/and quantitative blood flow measurements.

Acknowledgments

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