A Pipeline for Processing Ultrasound Image and Doppler Videos: Application to Dynamic Arterial Flowmetry and Diameter Measurement

Olivier Meste[1], Colin Lavigne[2], Fatimah Syam[1,3], Gregory M Blain[2]

Université Côte d'Azur, CNRS, I3S [1], LAMHESS [2], Nice, France
University Kuala Lumpur [3], Kuala Lumpur, Malaysia

Abstract

The objective of this study is to develop a novel method for enhancing the analysis of duplex ultrasound videos, enabling frame-by-frame measurements of blood flow and vessel diameter to better capture acute changes in these parameters.

In each frame, the B-mode and Doppler velocity components are processed separately based on pixel intensities and converted to cm/sec and mm, respectively, according to the scales included in the frames. The artery diameter is measured under the assumption that the two artery walls are parallel, approximated by two lines. Initially, a pseudo-optimal alignment is computed for the lower wall, followed by rotations and a secondary alignment for the upper wall. Doppler measurements are derived by referencing a specific column in the image and assuming the pixel intensities are modeled as a probability density function with a corresponding mean value.

Our approach for diameter measurement is confronted to an existing processing method from the literature, which has also been extended in this study. Under experimental conditions of reactive rest, hyperemia, and dynamic exercise, our method demonstrates superior robustness against tissue movement and improved performance for long-term acquisition, while enabling continuous tracking of the artery diameter.

1. Introduction

The assessment of human vascular function can be accomplished by measuring transient changes in blood flow and/or vessel diameter during various experimental conditions, such as reactive hyperemia (RH) and flow-mediated dilation (FMD) [1], passive leg movement (PLM) [2], or rapid onset vasodilation (ROV). These measurements can be non-invasively obtained through duplex ultrasound imaging and continuous Doppler velocity acquisition. Such evaluations provide prognostic insights into patients' cardiovascular health and fitness, while also offering valuable information on mechanisms related to exercise tolerance, blood pressure regulation, and vasomotor function.

The objective of this study was to utilize duplex ultrasound videos supplied by the equipment manufacturer to enable frame-by-frame measurements of blood flow and vessel diameter by non-experts, circumventing the need for radio frequency data processing. Furthermore, compared to existing methodologies [3], we introduce an observation model and processing techniques that enhance the robustness of vessel diameter measurements, addressing challenges related to the displacement, rotation, and vanishing of arterial walls during experimental protocols.

The effectiveness of the proposed processing pipeline is further demonstrated by the synchronization of diameter variation traces with blood velocity measurements, which can be computed with the highest temporal resolution.

2. Data acquisition and experimental protocols

Blood velocity and vessel diameter were measured with a (duplex ultrasound imaging device (ArtUs EXT-1H, Telemed Medical Systems, Vilnius, Lithuania). The ultrasound system was equipped with a linear array transducer (L12-5N40-A4, Telemed Medical Systems) and operating at a B-mode imaging frequency of 12 MHz and Doppler velocity frequency of 4 MHz using EchoWave II software (version 3.4.4, Telemed Medical Systems). Blood velocity measurements were obtained with the probe positioned to maintain an insonation angle of $\leq 60°$ to the axis of the vessel and the sample volume (range: 1–5 mm) centered and maximized according to artery size.)

Reactive Hyperemia (RH) and Flow-Mediated Dilation (FMD) [1]: In a supine position, a 9 cm blood pressure cuff was placed on the upper right leg, just above the knee. Baseline measurements were taken for 1 minute with an ultrasound probe positioned 10 cm proximal to the cuff. The cuff was then inflated to 250 mmHg for 5 minutes. Blood flow data from the superficial femoral artery was collected for at least 30 seconds before and 2 minutes after cuff release.

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Dynamic Knee-Extension Exercise : Participants, seated upright with their leg attached to a dynamic knee extension ergometer [2], performed 9 minutes of low-intensity knee extension exercise at 30% of peak power output, with a contraction rate per minute set at 60. Blood flow was continuously monitored from the femoral artery.

An example of frame #200 from the video captured by the duplex ultrasound imaging device at rest is presented in Fig. 1. The upper section displays the B-mode image, denoted Artery Region of Interest (AROI) for all frames, of the femoral artery under baseline conditions. Two Regions of Interest (ROIs) are applied to all frames, corresponding to the proximal (upper yellow box, denoted as ROI_p) and distal (lower yellow box, denoted as ROI_d) walls of the femoral artery. These ROIs will be manually defined and used to measure variations in the artery's diameter. The white markers seen in this section will be automatically removed to prevent interference with the measurement.

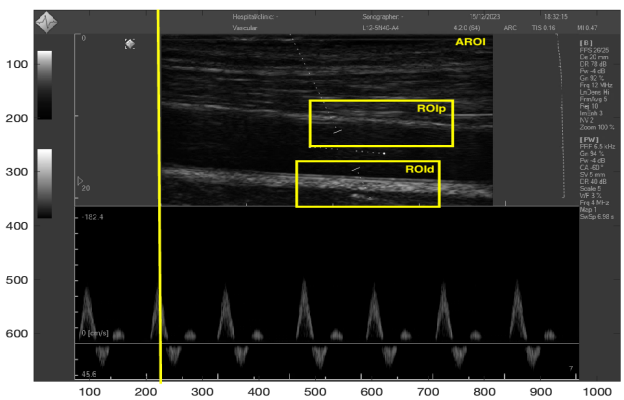

Figure 1. A frame (#200) recorded under baseline condition with a duplex ultrasound imaging and continuous Doppler velocity acquisition system. The two yellow boxes define ROI_p (upper) and ROI_d (lower) for the diameter measurement. The yellow line is considered as a reference column for the computation of the blood flow.

The lower section illustrates the blood velocity variations recorded by the continuous Doppler ultrasound system. The horizontal axis represents time, with pixel intensities indicating the velocity spectrum at each time point. In our case, negative values indicate that the blood flow is coming from the heart (systole). The third critical element of each frame includes the scales on the left side of the image, which will be interactively used to convert pixel indices to physical values during the final result display.

In each subsequent video frame, the ultrasound Doppler trace shifts leftward by a fixed number of pixels Nbpix, which is easily calculated ($Nbpix$=6 in this case), while a new image of the artery and surrounding tissues is captured. Synchronization between blood flow and artery diameter measurements is achieved by assuming that the current B-mode image corresponds to the Doppler spectrum displayed in the rightmost column of the lower section. It should be noted that this synchronization method may differ across acquisition systems.

The exploitation of the two main frame sections is described below.

3. Blood flow measurement

First, a reference column, denoted as $colref$, is selected from the Doppler ultrasound section of the frame f. An example of the corresponding pixels intensities $I_{f,colref}(p)$ along the yellow line in Fig. 1 is shown in Fig. 2. In this figure, the blood velocity spectrum is located within the horizontal range [500-600], and the zero-velocity line (0 cm/s) corresponds to the horizontal range [610-620], which will be replaced with "Not a Number" (NaN) values. Finally, for each frame f, the mean $m(f)$ of the spectrum will be computed over the interval [370-680] using the following formula:

$$m(f) = \sum_{p=370}^{p=680} pI_{f,colref}(p) / \sum_{p=370}^{p=680} I_{f,colref}(p) \qquad (1)$$

Because the Doppler ultrasound trace shifts leftward by a number of pixels equal to $Nbpix$ between two consecutive frames, the $Nbpix$ columns to the right of $colref$ can be similarly processed before analyzing frame $f+1$. This approach allows for a high-resolution computation of blood velocity, effectively enhancing the temporal resolution beyond the original frame rate of N frames per second. With the equipment used, the initial resolution or frame rate is 21 frames per second, resulting in a high-resolution blood flow measurement of 126 values per second. As the values in $m(f')$, with f' corresponding to the high-resolution index, are represented in pixels, they are converted to $\tilde{m}(f')$ (expressed in cm/sec) via a linear transformation, with the parameters derived from the scale provided on the left side of the image in Fig. 1. Note that for certain applications, the variance of the spectrum may also be calculated to provide insights into the distribution of blood flow velocities along the scan line.

4. Artery diameter measurements

In contrast to the previous section, the artery diameter will be estimated at the same rate as the video frame rate. However, interpolation may be applied to match the high resolution described earlier. Initially, the white markers in Fig.1 are manually identified and replaced with NaN values in each frame. If the markers are stable throughout the sequence, cancellation can be achieved using a Casorati Matrix, as suggested in [3].

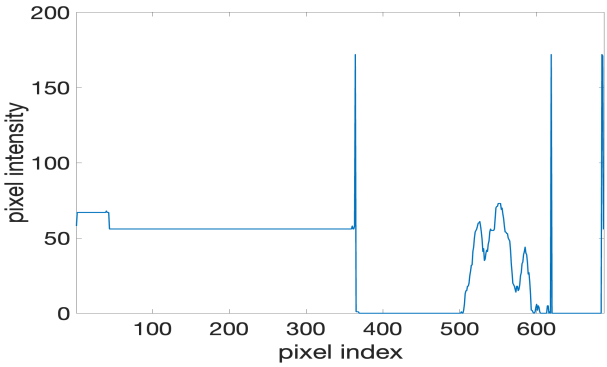

Figure 2. Pixels intensities along the reference column (yellow line in Fig. 1).

The proposed method assumes that, in each frame, the two artery walls are parallel and separated by a space known as the lumen, which varies from frame to frame. Additionally, the walls may rotate and shift vertically across frames due to movement of the ultrasound probe on the subject's skin or of the vessel (following muscle contraction for example). The processing pipeline for all the frames in AROI is as follows:

1. Frames subpixel alignment in AROI after estimating the vertical shifts of each frame in ROI$_d$(AROI) $\Longrightarrow$ AROI$_{s1}$
2. Frames subpixel alignment in AROI$_{s1}$ after estimating the rotations of each frame in ROI$_d$(AROI$_{s1}$) $\Longrightarrow$ AROI$_{s1r1}$
3. Frames horizontal alignment in the AROI$_{s1r1}$ following a unique rotation ROI$_d$(AROI$_{s1r1}$) $\Longrightarrow$ AROI$_{s1r1r2}$
4. Frames Subpixel estimation of vertical shifts in ROI$_p$(AROI$_{s1r1r2}$) $\Longrightarrow$ $l(f)$, representing lumen variations
5. Frames Subpixel alignment in AROI$_{s1r1r2}$ using $l(f)$ $\Longrightarrow$ AROI$_{s1r1r2s2}$
6. Manual measurement of the mean lumen value using the frames in ROI$_d$(AROI$_{s1r1r2}$) and ROI$_p$(AROI$_{s1r1r2s2}$) $\Longrightarrow d$
7. Computation of the variable lumen diameter in floating pixel values: $lu(f) = d + l(f)$
8. Conversion of $lu(f)$ to millimeters via a linear transformation, with parameters derived from the scale shown on the top left of the image in Fig. 1.

In this pipeline, steps 1 and 4 are extensions of the principle presented in [4] to two-dimensional data, where a vertical shift only is considered for each frame. The algorithm described in [4] is iterative and optimal, with a criterion based on the mean squared difference between the data to be aligned and the remaining dataset. The estimation of individual rotations in step 2 follows a similar approach, where the shift is replaced by a rotation. Step 3 involves minimizing the mean standard deviation across frames, evaluated horizontally for each rotated frame. This method utilizes a grid-search optimization to determine a single optimal rotation value for all frames.

The proposed approach is compared to the method presented in [3], denoted as $ro_1(f)$, where the varying lumen value is estimated differently, based on the difference between consecutive frames. To address the inherent limitations of the integration method used in [3], we introduce a slight modification by referencing the lumen value estimation solely to the first frame. This modification, denoted as $ro_2(f)$, eliminates the need for the aforementioned integration.

5. Results

In Fig. 3 it is observed that the peaks of $lu(f)$ are perfectly aligned with the nadir values of $\tilde{m}(f)$ during systole. This indicates that the femoral artery diameter increases as blood is pumped toward the legs. The mean arterial diameter is 7.2 mm. Additionally, the maximum negative peaks from the blood flow waveform (shown at the bottom of the figure 3) during systole can be used to calculate the superficial femoral artery blood flow per beat and per minute, yielding to 4 mL/beat and 224 mL/min in this example, respectively. The Under FMD condition in Fig. 4, where the cuff is inflated, we can observe a decrease of the absolute value of $\tilde{m}(f)$ and an increase of $lu(f)$ (mean value equal to 7.4 mm) compared to the baseline condition. In Fig. 5, it can be observed that blood velocity $\tilde{m}(f)$ abruptly increased and artery diameter $lu(f)$ decreased at cuff release followed by a slow return to the baseline condition values. Artifacts in the artery diameter are due to strong vanishing of the artery walls in the image. Figure 6 illustrates the effect of 30% dynamic knee extension, showing that the baseline of $\tilde{m}(f)$ is above zero, while the synchronization between $\tilde{m}(f)$ and $lu(f)$ is maintained.

6. Conclusions

This study demonstrates that duplex ultrasound imaging combined with continuous Doppler velocity video can be effectively utilized to accurately measure artery diameter and blood flow, even under conditions where the quality of the ultrasound image is suboptimal. A comparison between our original approach and existing methods in the literature shows that our method achieves high accuracy while preserving the synchronicity of measurements.

References

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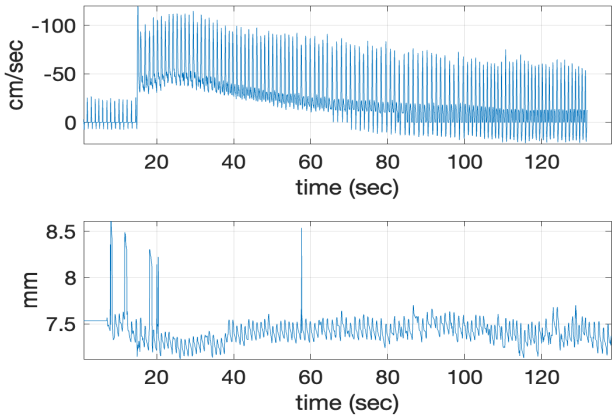

Figure 5. Subject 2. Estimated blood velocity (top, $\tilde{m}(f)$) and artery diameter (bottom, $lu(f)$) during the cuff release of the reactive hyperemia procedure at time $t = 15$ sec.

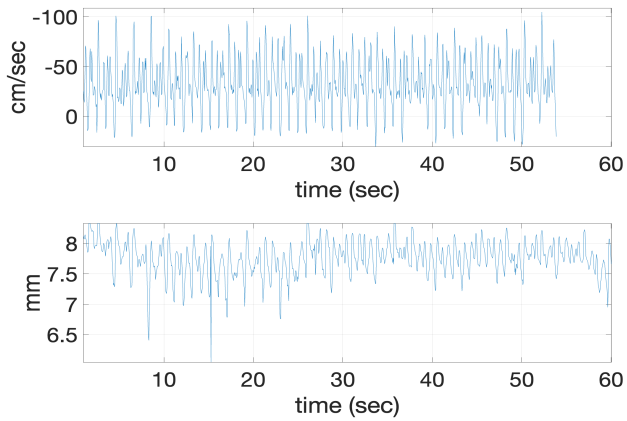

Figure 6. Subject 1. Estimated blood velocity (top, $\tilde{m}(f)$) and artery diameter (bottom, $lu(f)$) during dynamic knee-extension exercise for assessment of blood flow at 30% of peak power output.

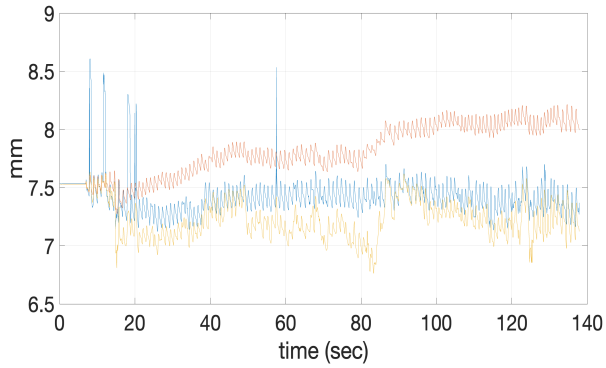

Figure 7. Subject 2. Comparison of the estimated artery diameter $lu(f)$ (blue), $ro_1(f)$ (red), $ro_2(f)$ (yellow), during the cuff release at time $t = 15$ sec.

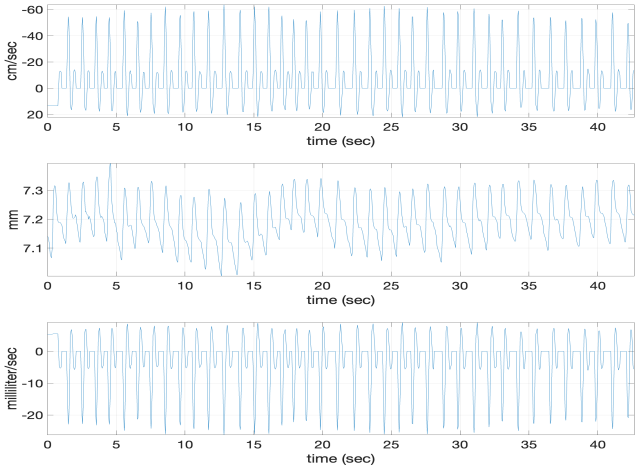

Figure 3. Subject 2. Estimated blood velocity (top, $\tilde{m}(f)$), artery diameter (middle, $lu(f)$) and the estimated blood flow (bottom) under baseline condition.

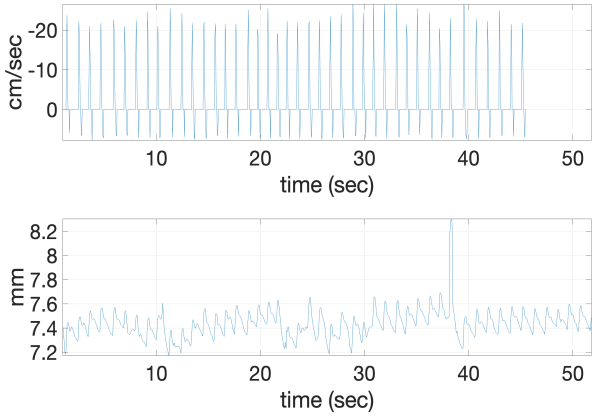

Figure 4. Subject 2. Estimated blood velocity (top, $\tilde{m}(f)$) and artery diameter (bottom, $lu(f)$) under FMD condition

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Address for correspondence :

Olivier MESTE, Laboratoire I3S - Université côte d'Azur - CNRS, 2000 route des Lucioles, 06903 Sophia Antipolis cedex, FRANCE. E-mail address: olivier.meste@univ-cotedazur.fr