Spectral broadening of ophthalmic arterial Doppler signals using STFT and wavelet transform

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Abstract

In this study, short-time Fourier transform (STFT) and wavelet transform (WT) were used for spectral analysis of ophthalmic arterial Doppler signals. Using these spectral analysis methods, the variations in the shape of the Doppler spectra as a function of time were presented in the form of sonograms in order to obtain medical information. These sonograms were then used to compare the applied methods in terms of their frequency resolution and the effects in determination of spectral broadening in the presence of ophthalmic artery stenosis. A qualitative improvement in the appearance of the sonograms obtained using the WT over the STFT was noticeable. Despite the qualitative improvement in the individual sonograms, no quantitative advantage in using the WT over the STFT for the determination of spectral broadening index was obtained due to the poorer variance of the wavelet transform-based spectral broadening index and the additional computational requirements of the wavelet transform.

Keywords: Doppler ultrasound; Short-time Fourier transform; Wavelet transform; Spectral broadening; Ophthalmic artery stenosis

1. Introduction

Doppler ultrasound is a noninvasive technique which is widely used in medicine for the assessment of blood flow in vessels. It may be used to estimate blood flow, to image regions of blood flow and to locate sites of arterial disease as well as flow characteristics and resistance of ophthalmic arteries [1–3]. Doppler systems are based on the principle that ultrasound, emitted by an ultrasonic transducer, is returned partially towards the transducer by the moving red blood cells, thereby inducing a shift...
in frequency proportional to the emitted frequency and the velocity along the ultrasound beam. The Doppler shift frequency $f_D$ is proportional to the speed of the moving targets:

$$f_D = \frac{2vf \cos \theta}{c},$$

(1)

where $v$ is the magnitude of the velocity of target, $f$ is the frequency of transmitted ultrasound, $c$ is the magnitude of the velocity of ultrasound in blood, and $\theta$ is the angle between ultrasonic beam and direction of motion. Since the scatterers within the ultrasound beam usually do not move at the same speed, a spectrum of Doppler frequencies will be observed. By using spectrum analysis techniques, the variations in the shape of the Doppler spectra as a function of time are presented in the form of sonograms in order to obtain medical information [1–3].

Doppler ultrasound is used clinically to assess stenosis in ophthalmic arteries [3,4]. The presence of stenosis may be indicated by disturbed flow distal to the site of stenosis and this causes a broadening of the spectrum of the Doppler signal around peak systole. Quantification of stenosis severity can be done using an index extracted from the Doppler spectrum such as the spectral broadening index (SBI) [1]. Kaluzynski and Palko [5] studied the behavior of SBI and other indices under different conditions for the spectrum analysis of simulated Doppler signals and concluded that the instability of the spectral estimates has only a limited effect on the indices derived from the spectrum. Keeton et al. [6] also used simulated Doppler signals and studied the robustness of Fourier-based and autoregressive-based SBI in noise. Keeton and Schlindwein [7] used clinical Doppler signals and studied the behavior of the SBI derived from spectra obtained using autoregressive (AR) modelling compared to that of SBI based on fast Fourier transform (FFT) analysis. They concluded that although AR had better spectral matching characteristics than the FFT approach, there was no significant improvement in the estimation of the SBI by using the AR technique even in the presence of noise.

A number of spectral estimation methods have recently been developed for Doppler ultrasonic signal processing [8–11]. The resolution of the spectral estimator used limits the detection sensitivity of disturbance-induced spectral broadening. In this study, ophthalmic arterial Doppler signals obtained from 52 subjects, 17 of them had suffered from ophthalmic artery stenosis, were examined by taking into consideration of their sonograms. Since flow in arteries is pulsatile and the red blood cells have a random spatial distribution, the Doppler signal is time-varying and random. Therefore, short-time Fourier transform (STFT) and wavelet transform (WT) were used for spectral analysis of the ophthalmic arterial Doppler signals. Using these spectral analysis methods, the time-dependent spectral distributions were visualized and detailed documentations of the Doppler signals were obtained. These methods were compared in terms of their frequency resolution and the effects in determination of spectral broadening in the presence of ophthalmic artery stenosis.

2. Materials and method

2.1. Hardware

The data acquisition system used in this study consisted of five blocks as illustrated in Fig. 1. These are 10 MHz ultrasonic transducer, analog Doppler unit (Diasonics Synergy color Doppler
2.2. Spectral analysis of ophthalmic arterial Doppler signals

In this study, ophthalmic arterial Doppler signals were obtained from 52 subjects. The subjects were 28 females and 24 males aged 21–64 years (mean age of 30.5 ± 0.5 years). According to the examination results, 17 of 52 subjects had ophthalmic artery stenosis (10 females and 7 males with a mean age 35.5 ± 0.5 years) and the rest of them were healthy subjects (18 females and 17 males with a mean age 29.0 ± 0.5 years).

Doppler signal is conventionally interpreted by analyzing its spectral content. The STFT and the WT were selected to obtain sonograms which represent the changes in Doppler frequency with respect to time. MATLAB software package was used to form sonograms of the ophthalmic arterial Doppler signals. Diagnosis and disease monitoring are assessed by analysis of spectral shape and parameters [1–3,5–7]. The SBI has been used as the parameter traditionally associated with the measurement of flow disturbances that occur with stenosis [1,5–7]. In this study, the SBI was calculated from both the STFT and the WT-based sonograms. The SBI used in this study is defined as

\[
SBI = \frac{f_{\text{max}} - f_{\text{mean}}}{f_{\text{max}}},
\]

where \( f_{\text{max}} \) is the maximum frequency at peak systole and \( f_{\text{mean}} \) is the mean frequency at peak systole (Fig. 2).

2.2.1. Short-time Fourier transform

Spectral analysis of the Doppler signal is performed using the STFT, in which the signal is divided into small sequential or overlapping data frames and FFT applied to each one. The output of successive STFTs can provide a time-frequency representation of the signal. To accomplish this the signal is truncated into short data frames by multiplying it by a window so that the modified signal is zero outside the data frame. In order to analyze the whole signal, the window is translated in time and then reapplied to the signal.
In STFT analysis, the signal is multiplied by a window function \( w(t) \) and the spectrum of this signal frame is calculated using the Fourier transform. Thus

\[
\text{STFT}(t, f) = \left| \int_{-\infty}^{+\infty} x(\tau) w(\tau - t) e^{-j2\pi f\tau} d\tau \right|^2,
\]

where \( x(t) \) represents the analyzed signal.

In this study, the STFT was performed using a 64 point Hamming window with an overlap of 50% and no zero padding was used. The problem with the STFT is that both time and frequency resolutions of the transform are fixed over the entire time-frequency plane. In addition, choosing a short analysis window may cause poor frequency resolution. On the other hand, while a long analysis window may improve frequency resolution, it compromises the assumption of stationarity within the window. A more flexible approach would be to use a scalable window: a compressed window for analyzing high-frequency detail and a dilated window for uncovering low-frequency trends within the signal [8–10].

2.2.2. Wavelet transform

The WT addresses the problem of fixed resolution by using base functions that can be scaled. The wavelets act in a similar way to the windowed complex exponentials that are used in the STFT, except that with the WT the length of signal being analyzed is not fixed. It is known that wavelets are better suited to analyzing nonstationary signals, since they are well localized in time and frequency. The property of time and frequency localization is known as compact support and is one of the most attractive features of the WT. The WT of a signal is the decomposition of the signal over a set of functions obtained after dilatation and translation of an analyzing wavelet. The main advantage of the WT is that it has a varying window size, being broad at low frequencies and narrow at high frequencies, thus leading to an optimal time-frequency resolution in all frequency ranges. Furthermore, owing to the fact that windows are adapted to the transients of each scale, wavelets lack of the requirement of stationarity. Therefore, the WT has become a powerful alternative to the STFT in analysis of the Doppler signals.
Continuous wavelet transform (CWT) is defined by

\[ \text{CWT}(a,b) = \int_{-\infty}^{+\infty} x(t) \psi_{a,b}^*(t) \, dt, \]  

(4)

where \( x(t) \) represents the analyzed signal, \( a \) and \( b \) represent the scaling factor (dilatation/compression coefficient) and the time (shifting coefficient), respectively, and the superscript asterisk denotes the complex conjugation. \( \psi_{a,b}(\cdot) \) is obtained by scaling the wavelet at time \( b \) and scale \( a \):

\[ \psi_{a,b}(t) = \frac{1}{\sqrt{|a|}} \psi \left( \frac{t - b}{a} \right), \]  

(5)

where \( \psi(t) \) represents the wavelet.

In this study, the Morlet wavelet was used and it is defined as [9–11]

\[ \psi(t) = e^{(-t^2/2)}e^{j2\pi ft}. \]  

(6)

3. Results and discussion

The Doppler shift signal contains a wealth of information about blood flow occurring within the sample volume of the Doppler ultrasonography. The most complete way to display this information is to perform spectral analysis and present the results in the form of a sonogram. The variation in the shape of Doppler power spectrum as a function of time can be presented in the form of a sonogram. In sonograms, time is plotted along the horizontal axis, frequency along the vertical axis and the power at a particular frequency and time as the intensity of the corresponding pixel.

Sonograms show the periodic heartbeats and within each beat it is possible to visualize the systolic and diastolic flow as the heart contracts and then relaxes.

In Fig. 3, ophthalmic artery flow sonograms recorded from 29-year-old healthy subject (subject no: 2) are presented. Ophthalmic arterial Doppler sonograms in Figs. 3(a), (b) are obtained using the STFT and the WT, respectively. In Fig. 4, ophthalmic artery flow sonograms recorded from 33-year-old unhealthy subject having ophthalmic artery stenosis (subject no: 5) are given. Ophthalmic arterial Doppler sonogram in Fig. 4(a) is obtained using the STFT, in Fig. 4(b) using the WT. As it is seen from Figs. 3 and 4, there is a distinct qualitative improvement in the sonograms obtained using the WT over the STFT. In the STFT analysis, taking the FFT of a short data frame of the Doppler signal leads to a distortion of the spectral estimate and leakage of signal energy into spurious side lobes due to the sharp truncation of the signal. Therefore, the STFT sonograms have spurious frequencies and the STFT does not produce clear sonograms. The advantage of the WT over the STFT is the optimization of the time-frequency resolution.

The sonograms recorded from normal ophthalmic artery (Fig. 3) exhibit a clear window under the systolic peak, resulting from a relatively flat velocity profile at this point in the cardiac cycle, whilst those from stenosed ophthalmic artery (Fig. 4) show a degree of spectral broadening which is believed to be the result of disturbed nonaxial flow. A number of parameters related to the blood flow may be extracted from the sonogram and these are of high clinical value. One of the parameters derived from the sonograms is the SBI which is used for quantification of stenosis severity in arteries.
Fig. 3. Ophthalmic arterial Doppler sonograms recorded from 29-year-old healthy subject (subject no: 2): (a) STFT, (b) WT.
Fig. 4. Ophthalmic arterial Doppler sonograms recorded from 33-year-old unhealthy subject having ophthalmic artery stenosis (subject no: 5): (a) STFT, (b) WT.
In this study, the SBI values calculated from the STFT-based sonograms and WT-based sonograms were referred to STFT–SBI and WT–SBI, respectively. There are two approaches for producing an estimation of the average value of a parameter derived from the sonograms. The first is averaging spectra from different heartbeats, and then calculating the parameter from the averaged spectrum; the second is calculating the parameter from the individual spectra, and only then averaging the values obtained. In this study, the second method was used to estimate the SBI in order to avoid the inherent difficulties of time alignment of the first technique and the possibility of smearing the averaged spectrum if the alignment is not perfect.

The SBI value calculated for a particular heartbeat was averaged over a number of heartbeats in order to obtain a statistically valid SBI. In this study, a statistically significant value for SBI was obtained using five heartbeats for a particular subject. The magnitude of the STFT–SBI and the WT–SBI for 10 subjects having ophthalmic artery stenosis is illustrated in Fig. 5. As it is seen from Fig. 5, the magnitude of the WT–SBI is significantly smaller than that of the STFT–SBI. The correlation between the SBIs obtained using the STFT and WT-based sonograms for 10 subjects having ophthalmic artery stenosis was calculated with the use of statistical tools such as correlation coefficients ($r$). The correlation coefficient between the STFT–SBI and the WT–SBI was calculated with a statistical package (SPSS version 8.0). The correlation coefficient is limited with the range $[-1, 1]$. When $r = -1$ there is a perfect positive linear correlation between the STFT–SBI and the WT–SBI, which means that they vary by the same amount. When $r = 1$ there is a perfectly linear negative correlation between the STFT–SBI and the WT–SBI, that means they vary in opposite ways. When $r = 0$ there is no correlation between the STFT–SBI and the WT–SBI. Intermediate values describe partial correlations. The calculated correlation coefficient between the STFT–SBI and the WT–SBI for 10 subjects having ophthalmic artery stenosis was equal to 0.837 that indicated there was perfect positive linear correlation between the STFT–SBI and the WT–SBI. The variance of the STFT–SBI and the WT–SBI for 10 subjects having ophthalmic artery stenosis is given in Fig. 6. From Fig. 6 one can see that the variance of the STFT–SBI is smaller than that of the WT–SBI. Therefore, the SBIs calculated using the STFT-based sonograms are generally more stable.
4. Conclusion

Doppler sonograms of the ophthalmic arterial Doppler signals were obtained using the STFT and the WT. These methods were compared in terms of their frequency resolution and the effects in determination of spectral broadening in the presence of ophthalmic artery stenosis. Based on the results obtained for the two techniques, it is clear that the WT can help improve the quality of the sonogram of the ophthalmic arterial Doppler signals. The WT do not suffer from some of the intrinsic problems that affect the STFT and hence there is a distinct qualitative improvement in the visualisation of the WT sonograms over the STFT sonograms. The SBI was calculated from both the STFT and the WT-based sonograms. A strong correlation was observed between the value of the STFT–SBI and the WT–SBI. The calculated STFT–SBI is larger than that of the WT–SBI and the variance of the STFT–SBI is smaller than that of the WT–SBI. The results of this study has shown that despite the qualitative improvement in the individual sonograms, there is no quantitative advantage in using the WT over the STFT for the determination of SBI due to its poorer variance and the additional computational requirements.

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References


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