Characterization of arterial stenosis and elasticity by analysis of high-frequency pressure wave components

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Abstract

We describe a novel method for assessing stenosis severity based on pressure wave measurements. Pressure waves for several degrees of stenosis at different distances proximal to the stenosis were recorded from in vitro and in vivo models. Signal analysis was performed using power spectral density, and radial compliance was also measured. Pressure wave components at the acoustic frequency band (400–2500 Hz) changed gradually and were dependent upon the distance from the stenosis and its severity. The shift of the pressure components could also demonstrate the elastic properties of tubes and arteries and explain the effect of a bifurcation in the system.

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1. Introduction

The severity of vascular stenoses can be estimated by measuring intravascular pressure and flow [1]. The development of miniature ultrasound and pressure probes enabled their insertion into the arteries in close proximity to a stenosed area without significantly affecting the hemodynamic behavior and accuracy of the measurements [2].

The problems that are encountered in assessing the extent of a stenosis stem partly from the non-linear relationship between the degree of the stenosis and the rate of blood flow and pressure...
This association can be further complicated by discontinuities in geometrical features and structural characteristics of the vessels, such as angulations and bifurcations, which are responsible for pressure and flow wave reflections. The currently available methods are also limited in their ability to accurately determine the degree and delineate the functional significance of relatively small stenoses.

There are currently several approaches for carrying out intravascular measurements, among them coronary flow reserve, hyperemic flow versus pressure slope index, fractional flow reserve, coronary flow velocity reserve and myocardial fractional flow reserve [4–7]. These methods generally require precise localization of the probe relative to the stenosis as well as the use of vasodilator drugs.

Several studies have analyzed the forward and backward waves in an arterial system in order to distinguish between physiological and pathophysiological conditions [8]. Stergiopulos et al. [9] found that a stenosis can induce wave reflection, even when the stenosis was hemodynamically not severe and angiographically non-detectable. Thus, the analysis of reflected waves may have important clinical and pathological significance.

The distance from the stenosis ($L$) and its severity ($\%S$) determine the components of the pressure wave that are measured proximal to the stenosis. It follows that an analysis of pressure and flow waveforms in different locations relative to the stenosed area could allow the quantification of the severity of an arterial stenosis.

Akay et al. [10,11] found that high-frequency acoustical power between 200 and 800 Hz was associated with turbulence produced by the partially occluded femoral arteries of the dog. They further noted that there is a correlation between the degree of the arterial stenoses and the energy contained in high-frequency bands. These findings led to a new approach that focused upon high-frequency bands, following dissipation of the harmonics of the heart [11]. Additionally, the turbulence of the flow creates vortices behind the stenosis. Vortex-shedding components are reflected from the vortices [12] and add to the components that are reflected from the stenosis, whereupon they affect the high-frequency components.

A numerical model which can characterize the stenosis by spectral analysis of pressure and flow waves was developed by us and others [13,14]. In order to perform a complete and accurate assessment of wave reflections, there must be an analysis of both the location and the severity of a stenosis [11]. Computer simulations have been used in the past to define the location of the occlusion [13,15].

The elastic properties of the artery change and, as a result, there are changes in the characteristics of the atherosclerosis due to increasing hydrodynamic wall shear forces and stretch. Thus, an understanding of the dynamic elastic behavior of the coronary arteries is of major importance for characterizing a stenosis and its elastic properties.

The elastic properties of the artery change with age and, as a consequence, so do those of arteries with atherosclerosis and calcification. Elasticity of arteries may have clinical importance, for example, for the choice of a suitable interventional device and for the development of improved arterial grafts [16]. Diverse methods have been suggested to obtain the values of arterial compliance [17], but none has been universally accepted [18].

Early detection of a coronary occlusion is of vital importance in terms of patient management and of clinical significance in terms of prognosis. To the best of our knowledge, this is the first report of an experimental study that uses a minimally invasive method to examine arterial pressure and flow waves and their spectrum frequency modulation resulting from arterial stenosis.
2. Methods

2.1. The in vitro closed system (Fig. 1)

The unit displayed in Fig. 1 consists of an electronically controlled pulsatile flow pump that operates at a rate of 70 cycles/min, pumping saline solution through latex tubes which have an external diameter ($R_o$) of 5 mm and an internal diameter ($R_i$) of 3 mm. It has parallel branches, a flow regulator and a reservoir with saline and air (for compliance adjustment) which control and calibrate the saline flow throughout the measured segment at 80 ml/min and at a pressure of 120 mmHg during the injection period and at 80 mmHg during the filling period. We matched the output impedance of the pump to the input impedance of the tubes. The in vitro experiments were performed in three different tubes:

1. A latex tube 30 cm long, with an $R_o$ of 3.45 mm and an $R_i$ of 2.75 mm.
2. A rubber tube 125 cm long, with an $R_o$ of 3.35 mm and an $R_i$ of 2.15 mm.
3. A clinically used graft (Hybrid PTFE™ Vascular Grafts, Atrium, NH, USA) 40 cm long, with an $R_o$ of 3.7 mm and an $R_i$ of 3 mm.

![Fig. 1. A scheme of the in vitro system. The arrows show the direction of flow. US indicates ultrasonic.](image-url)
An ultrasonic flowmeter measured the flow (Transonic System, Inc., Model T206, New York, USA). An ultrasonic flowmeter probe with a 4 mm $R_i$ (4NRB) was affixed upon the system, external to the tube and 45 cm proximal to the stenosis. Controlled stenoses were induced in the measured segment using an external clamp. Pressure catheters 2.5F in diameter (SPR-524, Millar Instrument, Inc., Texas, USA, frequency response flat up to 10 KHz) were inserted into the system in the direction of the flow using a Piton™ Tri-Adaptor (AC4002P, Medtronic, Israel) with a homeostatic valve for intra-tube pressure measurement.

2.2. *In vivo* experiments

Experiments were performed on 8 mongrel dogs (average weight 30 kg). General anesthesia was induced with intramuscular diazepam 1 mg/kg and intravenous thiopentone 6 mg/kg. Each dog was intubated and ventilated by a Harvard respirator. General anesthesia was maintained with halothane 1.5–2.5%. Electrodes were attached to the animal’s limbs, and an electrocardiogram was continuously recorded on a polygraph (for control purpose, to be sure that there is no arrhythmias). The femoral arteries were exposed from the groin to the knee, and all the branches were tied. The ultrasonic flowmeter probes with an $R_i$ of either 2 or 4 mm (2SB or 4SB) were placed on the external surface of the femoral artery close to the boundaries of the exposed segment, proximal and distal to the stenosis. To measure intra-arterial blood pressure, a Millar pressure catheter was inserted and positioned proximal to the stenosis, along the blood flow direction (Fig. 2). The left carotid artery was exposed for the control of systolic and diastolic blood pressure measurements.

A controlled stenosis was induced in the left femoral artery, adjacent and proximal to the distal flowmeter probe, using an external balloon catheter (IVM, vascular occluder, 0C4). Following the last measurement in the left femoral artery, the same procedures and measurements were carried out in the right femoral artery.

At the end of the measurements, the dog was euthanized by an intravenous overdose of sodium pentobarbital and potassium chloride. The investigation conformed with the guide for the care and

![Fig. 2. A scheme of the femoral artery dog segment. US indicates ultrasonic.](image-url)
use of laboratory animals of the US National Institutes of Health (NIH publication No. 85-23, revised 1996).

2.3. The protocol

In vitro experiments: Pressures and flows were recorded at varying distances between the pressure catheter tip and the stenosis: at 5, 10, 15, 20 and 25 cm for the latex tube and the graft, and at those distances as well as at 30 and 40 cm for the rubber tube.

A series of recordings was made at each point during varying degrees of stenosis, causing a reduction in blood flow of 0%, 25%, 50%, 75% and 99% of the initial value.

In vivo experiments: The experiments were performed in the animal’s femoral artery. The pressures and flows were recorded at the above-mentioned occlusion percentages, but at different distances between the pressure catheter tip and the stenosis (i.e., 4, 8 and 10 cm), due to the smaller length of the exposed arteries.

2.4. The sampling and data recording system

Hardware: The variables were continuously recorded during the in vitro experiments on a PC-based frame, using a 12-bit A/D conversion. A short circuit channel was placed between the pressure and flow channels to prevent reciprocal influence. The sampling rate was 5000 Hz, and the dynamic ranges were ±500 mV for most of the experiments. The input range was increased to ±1.0 V in both the latex tube and in the graft for high degrees of occlusion (75% and 99%).

Software: The signal sampling and the transmission to the computer were performed using DasyLab software (DasyLab®, DASYTEC, Germany). Signal processing was performed and computer programs were written by means of MATLAB software (MATLAB®, The MathWorks Inc., USA).

2.5. Signal analysis

One section without interference was selected for each stage of the experiment.

2.5.1. Research FFT components in different frequency bands

Power spectral density was performed on this section using Welch’s average periodogram method with a Hanning window fixed at 16284 sampling points, and was computed from non-overlapping segments of windowed intervals:

1. A logarithmic scale was used for low-frequency bands (0–20 Hz).
2. The various stages were integrated.
3. A thorough search was performed throughout all frequencies, i.e., 0–2500 Hz. (Note: 2500 Hz is the highest limit according to Nyquist law).
4. The local maximum of the largest power density of the pressure components, at each stage in varying frequency bands, was determined by using a computer program written specifically for this purpose.
5. Fitting of linear and second-order polynomials of the maximal pressure components versus the distance from the stenosis and its degree was carried out.
2.6. Wave velocity measurements

(1) The time delay between the pressure signals that were recorded at two different locations (with distance of 20 cm) was determined.
(2) The wave velocity was calculated.

2.7. Elasticity properties measurements

Changes in the internal pressure (mmHg) against changes in the external radius (mm) were measured using a microscope (Toolmakers, Mitutoyo, Japan). The radial compliance \( C_r \) was calculated based on the conventional formula:

\[
C_r = \frac{R - D}{D_o},
\]

where \( D \) is the final external diameter, \( D_o \) is the initial external diameter and \( P \) is the internal pressure.

2.8. Statistical analysis

Analysis of variance was performed to compare the polynomial coefficients in the different tubes. A correlation test was used to examine the relation between the compliance of tubes and the polynomial coefficients. Statistical significance was defined as \( P < 0.05 \).

3. Results

3.1. In vitro experiments

The in vitro results showed that both pressure and flow waves changed with increasing degrees of stenosis in the high-frequency bands (Fig. 3).

High-frequency pressure wave components that were obtained from measurements in the latex tube (Fig. 4) focused upon two frequency bands, 700–900 Hz and 1540–1580 Hz. The \( X \)-axis indicates the stage of the experiments, which depends on the distance from the stenosis \( L \) and its severity \( %S \). The points between two distances indicate \( %S \) (0%, 25%, 50%, 75% and 99% stenosis). The \( Y \)-axis indicates frequency in Hz. It was clearly illustrated that the pressure components change gradually and that they were dependent both on the distance from the stenosis and on its severity. These diagonals appeared from 700–1600 Hz. Fig. 4 illustrates typical examples. In order to quantitatively characterize the changes, polynomials of the pressure components were fitted using a computer program written specifically for this purpose. The continuous lines describe the second-order polynomial functions that were fitted for the graphs. All the polynomials had an excellent correlation to these data \( (R^2 > 0.94) \). \( X \) in the polynomial could be calculated using Eq (1).

\[
X = L + \left( \frac{\%S}{25} \right) - 4.
\]  

(1)

The figures displayed a linear trend up to 20 cm from the stenosis and curved thereafter at the last measuring point (25 cm from the stenosis). This was attributed to the proximity of the triple
connector which reflected the waves that were most significant in this region. In order to examine the influence of \( L \) and \( \%S \) alone and to eliminate the effect of the connector, linear polynomial functions were fitted for the data pertaining to the section up to 20 cm from the stenosis. Curves A–B in Fig. 5 were comparable to graphs A–B in Fig. 4: there was a very strong linear relationship between the pressure components and \( X \) for both curves \( (R^2 > 0.98) \).

Fig. 6 demonstrates that the behavior of pressure components in the high-frequency bands was similar to that of a latex tube during the experiments with a clinically used graft. The focus in this figure was on the bands 500–1200 Hz and 1500–1600 Hz, while A and B are the graphs of the pressure wave components including the second-order polynomial functions that were fitted \( (R^2 > 95) \). The data points for each polynomial are very clearly observed as forming 5 separate groups. Each group indicates the wave components that were obtained from distances from the stenosis of 5, 10, 15, 20 and 25 cm. It can also be observed in this figure that the pressure components changed gradually and were dependent upon the distance from the stenosis, its severity, and the curve of the polynomial near the connector. The linear functions of graphs A and B in Fig. 7 were comparable to the data of graphs A and B in Fig. 6, respectively, up to \( L = 20 \) cm. A linear relationship between the pressure components and the \( L \) and \( \%S \) was also very clear \( (R^2 > 0.94) \).
Fig. 4. (A,B) The polynomial functions that fitted for pressure wave components obtained from an experiment with a latex tube. The X-axis indicates the stage of the experiment, depends on distances from the stenosis (L), and its severity (%S). The Y-axis indicates frequency in Hz ($R^2 = 0.94$).

frequency band of the upper figure (500–1200 Hz) was the lowest band for which this phenomenon appeared.

A linear correlation between the pressure wave components and L and %S also appeared during the experiments with a rubber tube (Fig. 8, $R^2 > 0.94$). However, due to the high velocity wave in this tube, the changes occurred from $L = 15$ cm onward. Technically, the tube was very long, enabling measurements up to $L = 40$ cm, while the tip catheter was still far from the connector. This is the reason that the line did not curve even at the furthermost point of measurement. The two linear curves were obtained, however, in frequency bands higher than 1500 Hz.

3.1.1. Compliance measurements

Radial compliance ($C_r$) was calculated as being the slope of the line fitted for changes in the external radius versus changes in the internal pressure of the latex tube. No change will appear
in the external radius of the tube ($\Delta r = 0$) when no change occurs in the internal pressure of the tube ($\Delta P = 0$). Therefore, the linear polynomial functions were fitted for the data by setting the $y$-intercept to zero. A comparison between the radial compliance ($C_r$) of the different tubes was performed and the results are summarized in Table 1. A significant difference was found between the $C_r$ values for all the tubes, but the difference between the latex tube and the graft was less significant ($P < 0.05$) compared with the difference between them and the rubber tube ($P < 0.001$).

3.1.2. Pressure wave velocity

The results of the pressure wave velocity ($V_w$) in different tubes are presented in Table 2. Each value is an average of three measurements. A significant difference was found between the $V_w$ values for all the experimental tubes ($P < 0.05$).

3.2. In vivo experiments

The pressure components also changed gradually and were dependent upon the distance from the stenosis and its severity during the animal experiments. Fig. 9 presents a typical example, focusing on the two frequency bands of 900–950 Hz and 2000–2300 Hz. The $X$-axis indicates the stage of the experiments, which depends on the distance from the stenosis ($L$) and its severity ($%S$). The points between two distances indicate $%S$. The $Y$-axis indicates frequency in Hz. The continuous lines describe the linear polynomial functions that were fitted for the graphs. The linear relationship
Fig. 6. (A,B) The polynomial functions that fitted for pressure wave components obtained from an experiment with a graft. The $X$-axis indicates the stage of the experiment, depends on distances from the stenosis ($L$), and its severity ($\%S$). The $Y$-axis indicates frequency in Hz ($R^2 > 0.95$).

between the pressure components and $X$ was very strong for both curves ($R^2 > 0.88$). Fig. 10 presents the results from another canine experiment that focused on the 2390–2440 Hz band. Both the pressure wave component (Fig. 10A) and the polynomial function that was fitted for this graph ($R^2 = 0.85$) indicated that the polynomial curved only at the last measuring point. The reason for this is that the pressure catheter was too close to the branch of the artery at the last measuring point ($L = 10$ cm) and, therefore, the reflected waves from the branch were the most significant, similar to the phenomenon observed in the tube experiments. To examine only the influence of $L$ and $\%S$ and to eliminate the effect of the branch, linear polynomial functions were fitted for the data retrieved up to 8 cm from the stenosis. The linear relationship between the pressure components and $X$ was a strong one ($R^2 > 0.84$).
4. Discussion

The persistence of coronary disease as the number one killer in developed countries has emphasized the need for enhanced methodology for the functional assessment of coronary stenosis [19,20]. Serial demonstrations of qualitative and quantitative angiographic morphology before and after percutaneous transluminal coronary angioplasty were shown to be poor predictors of clinical outcome, especially for immediate and long-term prognosis [21]. Clinically relevant studies with in-laboratory physiology continue to support the view that the physiological variable is as important—if not more so—as the anatomic variable, the latter being the ultimate functional parameter in diagnosing a patient’s condition [22].

The present study describes a method for defining the function of an arterial stenosis based on the reflected pressure wave components at high-frequency bands after the harmonics of the pump or the heart are dissipated. Based on the findings of Akay et al.’s studies [10,11], it is assumed
that high-frequency band components result from and are associated with the turbulence produced by partial occlusion.

Pressure wave components at the acoustic frequency band 400–2500 Hz changed gradually and were dependent on the distance from the stenosis ($L$) and its severity ($\%S$) during all in vitro and in vivo experiments. These changes can be attributed to the waves that were reflected from the stenosed region. It is known that vortices are created around stenosed areas since blood flow becomes unstable and even turbulent in the vicinity of a stenosis. In the same manner, the reflected waveform changes its frequency and amplitude along the arterial lumen [23]. Information obtained at the acoustic frequency range between 400 and 2500 Hz could be matched with a Strouhal number, based upon the values of diameters and liquid velocities that were used in the experiments [24]. The stenosis and the vortices increased concomitantly. The changes in frequency throughout the experiments indicated that the recorded pressure signal had included the vortex-shedding components that had been reflected...
Table 1
Radial compliance for all tubes

<table>
<thead>
<tr>
<th>Tube/series</th>
<th>Rubber</th>
<th>Latex</th>
<th>Graft</th>
</tr>
</thead>
<tbody>
<tr>
<td>Series 1</td>
<td>0.00020</td>
<td>0.00015**</td>
<td>0.00006*</td>
</tr>
<tr>
<td>Series 2</td>
<td>0.00023</td>
<td>0.00012**</td>
<td>0.00003*</td>
</tr>
<tr>
<td>Series 3</td>
<td>0.00021</td>
<td>0.00011**</td>
<td>0.00007*</td>
</tr>
<tr>
<td>Mean</td>
<td>0.00021</td>
<td>0.00013**</td>
<td>0.00005*</td>
</tr>
<tr>
<td>SD</td>
<td>0.00002</td>
<td>0.00002</td>
<td>0.00002</td>
</tr>
</tbody>
</table>

(*) Indicates significant difference between the certain tube and the rubber tube \( (P < 0.001) \). (**) Indicates significant difference between the latex tube and the graft \( (P < 0.05) \). Each value represents one series of measurements (in 1 mmHg).

Table 2
Pressure wave velocity for all tubes

<table>
<thead>
<tr>
<th>Tube/( V_w )</th>
<th>Rubber</th>
<th>Latex</th>
<th>Graft</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean</td>
<td>15.55</td>
<td>43.48***</td>
<td>28.78*</td>
</tr>
<tr>
<td>SD</td>
<td>4.95</td>
<td>0.01</td>
<td>0.29</td>
</tr>
</tbody>
</table>

(*) Indicates a significant difference between the certain tube and the rubber tube \( (P < 0.001) \). (**) Indicates a significant difference between the latex tube and the graft \( (P < 0.05) \). Each value is an average of three measurements (in m/s).

from the vortices that were added to the components already being reflected from the stenosis [12]. In addition, the shifting in the wave components also occurred according to the length of time it took them to travel from the stenosis to the measuring point, and differed between various measuring locations. This shift in time is determined both by the wave speed and the distance traveled [25]. The linear polynomial functions that were fitted depict the changes in pressure wave components dependent on \%S and \( L \) during both the in vitro and the in vivo experiments.

The inverse linear relationship between the frequencies and \%S can be explained by the fact that the pressure wave had the highest velocity when it crossed from the stenosis to an open region. In addition, the artery/tube wall was extended in the region adjacent to the stenosis, based on Laplace law and Lame’s equation [23]. Thus, the more the degree of stenosis increased, the more the pressure in the stenosis region and the tension of the wall tube increased, the expansion of the wall near the stenosis was more pronounced and, consequently, the frequency of the pressure wave components reflected from the wall was reduced, in accordance with the Doppler effect.

An additional explanation can be based on the resonance theory. First, there is the effect of the transverse waves in the tube. As a result of tube expansion, its self-resonance frequency decreases. Thus, while the degree of stenosis increases and the tube expansion is larger, its self-resonance frequency decreases and the dominant frequency reflected from the wall also decreases [26]. Second, there is the effect of the longitudinal waves between the sensor and the stenosis. In all probability, the stenosis and the pressure catheter created a resonator. According to the acoustic theory, the self-resonance frequency of a resonator decreases to the extent that its boundary narrows. Hence, the dominant frequency decreases in parallel with the increase in the degree of stenosis.
The inverse linear relationship between the frequencies and $L$ can be attributed to the behavior of the resonator between the sensor and the stenosis. The resonator’s length increases in parallel to the amount of increase in the distance between these locations. Therefore, the self-resonance frequency decreases, causing the observed frequency shifting. The diagonals shown are the result of the combined increasing of the distance from the stenosis and of its severity.

The frequency shifting can also result by a combination of the above effects. However, further experiments, including continuous measuring of the wall dimensions and recording of the wave’s paths using a digital video camera during the experiments, would be needed to substantiate that possibility.

Porret et al. [27] recently reported that the oscillations of arterial blood flow could induce significant changes in the arterial diameter. The values of the wall dimensions and the wave velocity depend on the elastic properties of the conduit and wall thickness [23]. Moreover, the reflection coefficient of the wave is also related to the elastic properties of the vessel [9]. Indeed, varying results of the measurements of radial compliance ($C_r$) and wave velocity ($V_w$) were observed in the current study. As expected, the measurements of the $C_r$ and $V_w$ indicated that the elasticity of the rubber tube is significantly greater than that of the latex tube and the graft ($P < 0.0001$). The values
Fig. 10. (A) The polynomial functions that fitted for pressure wave components obtained from a canine experiment. The $X$-axis indicates the stage of the experiment, depends on distances from the stenosis ($L$), and its severity ($%S$). The $Y$-axis indicates frequency in Hz ($R^2 = 0.85$). (B) The polynomial function that was fitted for graph (A), up to $L = 8$ cm. The $X$-axis indicates the stage of the experiment, depends on $L$ and $%S$. The $Y$-axis indicates frequency in Hz ($R^2 = 0.84$).

of the latex tube and the graft are similar to those for the canine femoral and coronary arteries [23]. The results of the $C_r$ and $V_w$ evaluations clarify these observations, since both these variables affected the shifting of frequency [25].

In the rubber tube experiments, the diagonals began only at 15 cm from the stenosis, while the diagonals observed throughout the other experiments began from the first measurement point (i.e., 5 and 4 cm in the in vitro and in vivo experiments, respectively). This difference is due to the above-mentioned difference in the elastic properties of the rubber tube compared with the other two tubes. According to the acoustic theory, it may be possible that the first two measuring points in this tube did not produce a resonator and, consequently, neither the transverse waves nor the longitudinal wave effects were produced.

The graphs of the second-order polynomial function in the latex tube, the graft and part of the canine experiment depict a linear trend up to the last measuring point and the trend to curve thereafter. The curve appeared because, at this point, the pressure sensor was too close to the triple connector or to the artery branch during the in vitro and in vivo experiments, respectively. (Technically, the rubber tube was very long, so the tip catheter was still far from the connector.
even at the furthermost measuring point). This phenomenon is important since it demonstrates the behavior of the pressure signal close to the branches. The branches and bifurcation effect on the pressure and signal is substantial and was widely investigated [28]. These findings could contribute to elucidate the alterations of hemodynamic signals close to the branches.

4.1. Limitation of the study

The in vivo experiments were performed in the canine femoral artery since its dimensions and flow values are similar to those of human coronary arteries [11] and enable accurate and controlled measurements. Technically, however, the measurements of the various degrees of stenosis could only be taken at 2 or 3 distances from the stenosis due to the short length of the artery. Therefore, different distances between the pressure tip and the stenosis were used for the in vitro and in vivo experiments.

Direct measurement of the arterial compliance is difficult and frequently non-attainable [18]. Since the mechanical properties of the artery change after isolation [27,29,30], the compliance measurements and calculations were performed only for the tubes and the graft.

4.2. Clinical implication

This study describes a phenomenon, which is dealt with a new approach. Changes in the pressure wave components at high frequency band (acoustic waves) depend upon the distance from and severity of the stenosis as well as vessel material. The method described in this work was found to be sensitive to low degrees of stenosis, and could thus estimate the changes in the pressure waves at intermediate degrees of stenosis as well. Furthermore, it could contribute to clarifying the alteration of the hemodynamic signals adjacent to the branches of a vessel. Hence, analysis of pressure waves by the above-described method may potentially define the severity and localization of arterial stenosis and characterize the elasticity of the artery. In addition, the measurements are performed only proximally to the stenosis. The clinical significance of this procedure is clear, since there is little risk, by obtaining functional evaluation without inserting a probe through the stenosis. This study contributes to a better understanding of arterial function under normal and pathological conditions, and could lead to a more efficient functional assessment of arterial stenosis.

5. Summary

The method for pressure wave analysis described herein is a novel approach that is based on information concerning very high-frequency bands (400–2500 Hz) for assessing stenosis location and severity. The frequency bands in this study are suitable to those studied in the previous investigations based on this approach [11,13]. Pressure wave components at the acoustic frequency band changed gradually and were dependent upon the distance from the stenosis and its severity in both in vitro and in vivo experiments. The elastic properties of the tubes and arteries could be characterized by using the functions for demonstrating the shift at high frequencies. The polynomial curvature could explain the effect of a bifurcation on the pressure signal. It is our contention that this method could
lead to more effective diagnosis of functional stenosis in the clinical setting. The results can serve as a very good basis for future in vivo and clinical works.

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