An Ultrasound Doppler Technique for the Noninvasive Determination of the Pressure Gradient in the Björk-Shiley Mitral Valve

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SUMMARY The accuracy in determining the pressure gradient in the Björk-Shiley mitral implant from noninvasive ultrasound Doppler data was explored in nine adult patients. Manometric and ultrasound data were collected simultaneously, and identical diastolic periods were used to compare the manometric gradient (ΔP_M) with the gradient obtained from ultrasound data (ΔP_U). In the nine patients the mean diastolic value of ΔP_M ranged from 2–12.5 mm Hg and the difference between the mean diastolic values of ΔP_M and ΔP_U was 0.3 ± 1.0 mm Hg (SD). The results of the investigation indicated that the method is accurate and reliable in the noninvasive determination of the mean diastolic gradient in the Björk-Shiley mitral implant.

The pressure gradient in implanted mitral valves is frequently used in combination with the cardiac output to obtain a quantitative expression for the flow obstruction in the implants. Determination of the gradient requires simultaneous left atrial and left ventricular catheterization; a technique that can determine the gradient noninvasively would therefore be a valuable diagnostic tool.

Previous reports have described a noninvasive ultrasound Doppler technique that can determine the pressure gradient in mitral stenosis with an accuracy sufficient for diagnostic purposes.1,2 Noninvasive ultrasound Doppler data have also been used to evaluate the Björk-Shiley mitral implant. The results indicated that the gradient in the implant may be determined noninvasively.3,4 In this investigation we explore the accuracy of ultrasound in determining the gradient in the Björk-Shiley mitral implant.

Methods and Materials

Theoretical Considerations

In the steady flow through an orifice situated between two reservoirs, Torricelli's law (equation 1) allows accurate prediction of the pressure gradient if the Reynolds number is sufficiently large:

\[ \Delta P = \frac{1}{2} \rho V_{\text{MAX}}^2 \]  

where \( \Delta P \) = pressure gradient, \( \rho \) = mass density of fluid, and \( V_{\text{MAX}} \) = maximum fluid velocity.

Torricelli's law describes a quadratic relationship between \( \Delta P \) and \( V_{\text{MAX}} \) and allows the calculation of \( \Delta P \) if \( V_{\text{MAX}} \) is known. When the Reynolds number is smaller than that necessary for the validity of the law the application of equation 1 will underestimate the actual gradient.5

The diastolic flow through the Björk-Shiley mitral valve is similar to the flow through an orifice in that the sewing ring and the tilted disc constrict the available flow area between the left atrium and the left ventricle. In vitro studies of the flow through the valve have demonstrated a quadratic relationship between \( \Delta P \) and flow rate,6,7 and since the valve has a fixed geometric configuration in the open position, this indicates that Torricelli's law is valid for the flow through the valve. It may thus be possible to determine \( \Delta P \) noninvasively if \( V_{\text{MAX}} \) can be determined with ultrasound.

Noninvasive ultrasound Doppler instruments can obtain the frequency shifts (Doppler spectrum) due to the velocities of the corpuscles in blood streams within humans. In its simplest “display” mode the instrument presents the Doppler spectrum as an audio signal to the operator. The relationship between the velocity of an object and the frequency shift is expressed by the Doppler equation:

\[ V = \frac{c \cdot \Delta f}{2 \cdot f \cdot \cos \theta} \]  

where \( V \) = velocity of reflecting object, \( c \) = velocity of sound, \( f \) = frequency of incident sound beam, \( \Delta f \) = frequency shift, and \( \theta \) = angle between axis of incident sound beam and velocity vector of \( V \).

The Doppler equation states that if \( \Delta f \) and \( \theta \) are known, the magnitude of \( V \) can be determined. For a given \( V \) the largest \( \Delta f \) will be obtained if the incident sound beam travels directly along the path of the reflecting object (\( \theta = 0^\circ \), \( \cos \theta = 1 \)).

When Doppler instruments are used to obtain velocity information from a blood stream, the incident sound beam usually will be reflected from corpuscles moving with different velocities. The signal will thus contain a spectrum of frequency shifts. A frequency analysis of the spectrum will reveal information about each component frequency shift. The frequency analysis will also identify the largest frequency shift (\( \Delta f_{\text{MAX}} \)) in the spectrum at a given time; thus, the time course of \( \Delta f_{\text{MAX}} \) will be identified. If it is assumed that \( \Delta f_{\text{MAX}} \) is due to the corpuscles moving with the largest velocities (\( V_{\text{MAX}} \)) the Doppler equation will allow the determination of the time course of \( V_{\text{MAX}} \) if the angle \( \theta \) is known.

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Previous studies have demonstrated that a 2.1 MHz Doppler instrument can consistently obtain the Doppler spectrum from the diastolic flow through the Björk-Shiley mitral valve.\(^3\)\(^4\) Chest radiograms have shown that the axis of the sewing ring points toward the lateral and inferior precordium. Thus, it may be possible to obtain the Doppler spectrum from the flow through the valve with the ultrasound probe positioned on the precordium such that the axis of the incident sound beam coincides with the direction of the vectors of \(V_{\text{MAX}}\) (cos \(\Theta = 1\)).

In any patient the optimum probe position will theoretically result in a Doppler spectrum that contains the largest diastolic values of \(\Delta f_{\text{MAX}}\). The previous studies of patients with the Björk-Shiley valve demonstrated that the diastolic values of \(\Delta f_{\text{MAX}}\) are audible, so it may be possible to use the audio signal of the Doppler spectrum as a guide in a scanning procedure designed to identify the optimum probe position.

Assuming that \(\Delta f_{\text{MAX}}\) has been obtained with a probe position compatible with cos \(\Theta = 1\), the Doppler equation becomes

\[
V_{\text{MAX}} = \frac{c \cdot \Delta f_{\text{MAX}}}{2 \cdot f}
\]

Inserting the right hand side of equation 3 in equation 1 and using \(c = 1570 \times 10^3 \text{ cm/s}\), \(\rho = 1.06/\text{g/cm}^3\), and \(f = 2.1 \text{ MHz}\) yields the following working equation:

\[
\Delta P_u = 0.55 \cdot \Delta f_{\text{MAX}}^2
\]

where \(\Delta P_u\) = pressure gradient as determined with ultrasound (mm Hg) and \(\Delta f_{\text{MAX}}\) = maximum frequency shift (kHz).

**Patient Material**

Nine adult patients with a Björk-Shiley mitral valve were studied (table 1). Valve replacement had been performed 12–24 months before the investigation and the valves were implanted with the larger opening posteriorly.

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**Ultrasound System**

A modified\(^6\) 2.1 MHz non-directional continuous waveform Hewlett-Packard Sound Monitor was used to obtain the Doppler spectrum from the diastolic mitral blood flow. The spectrum was recorded on magnetic tape and frequency analyzed on a Kay Sonagraph Sound Spectrograph. The hard copy of the frequency analysis is a gray-scaled presentation where the amount of blackening is related to the energy at each component frequency (figs. 1–4).

**Manometric System**

Left atrial and left ventricular pressures were obtained via #7F catheters (transseptal left atrial and retrograde left ventricular catheterization). The catheters were connected to EMT-34 transducers (Elema Schönander, Stockholm) and pressures registered simultaneously by a Mingograph 800 (Elema Schönander).

**Data Collection**

Ultrasound data and manometric data were collected simultaneously with the patient resting in the supine position on the catheterization table. Before data collection the operator of the ultrasound instrument scanned the region of the mitral valve with the ultrasound beam from several positions on the precordium and noted the region where the audio signal of the Doppler spectrum had the highest frequencies. A subsequent, more meticulous scan was then performed with the probe in this region and data collection begun when the apparent optimum probe position had been identified. Data from 10–20 consecutive cardiac cycles were thus collected.

After collecting ultrasound and manometric data, cardiac output determination and left ventricular angiography were performed.

**Construction of Gradients**

Three or four consecutive cardiac cycles which had a minimum of background noise were selected for frequency analysis. The curve representing \(\Delta f_{\text{MAX}}\) was drawn by hand on the hard copy of the frequency

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**Table 1. Patient Data and Numerical Results (Gradients Rounded off to Nearest 0.5 mm Hg)**

<table>
<thead>
<tr>
<th>Pt</th>
<th>Age (yr)</th>
<th>Sex</th>
<th>Valve size</th>
<th>R</th>
<th>HR</th>
<th>(\dot{Q})</th>
<th>(\Delta P_M)</th>
<th>(\Delta P_P)</th>
<th>Angiography</th>
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<td>1</td>
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<td>F</td>
<td>AF</td>
<td>29</td>
<td>64</td>
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<td>MI</td>
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<tr>
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<td>52</td>
<td>F</td>
<td>AF</td>
<td>29</td>
<td>92</td>
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<td>3.0</td>
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<tr>
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<td>AF</td>
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<td>8.0</td>
<td>6.5</td>
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</tr>
<tr>
<td>4</td>
<td>55</td>
<td>M</td>
<td>AF</td>
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<td>12.5</td>
<td>11.0</td>
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<tr>
<td>5</td>
<td>56</td>
<td>F</td>
<td>AF</td>
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<td>85</td>
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<td>5.5</td>
<td>6.0</td>
<td></td>
</tr>
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<td>6</td>
<td>48</td>
<td>M</td>
<td>AF</td>
<td>31</td>
<td>64</td>
<td>5.8</td>
<td>6.0</td>
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**Abbreviations:** R = rhythm; HR = heart rate; \(\dot{Q}\) = cardiac output (l/min); \(\Delta P_M\) and \(\Delta P_P\) = manometric and ultrasound mean diastolic gradient, respectively (mm Hg); MI = mitral insufficiency; AI = aortic insufficiency; AF = atrial fibrillation; SR = sinus rhythm.
analysis as a line enveloping the blackened areas representing the frequency shifts of the diastolic blood flow. Discrete points on this curve were then inserted in equation 4 and the values of \( \Delta P_U \) used to construct the time course of \( \Delta P_U \). The time course of \( \Delta P_M \) was constructed from discrete values obtained by subtracting the left ventricular pressure from the left atrial pressure. Since the hard copy of the frequency analyses did not have an ECG tracing, the alignment in time of the constructed gradients was arbitrary in figures 1–4; the leading edge of \( \Delta P_U \) was simply aligned with the leading edge of \( \Delta P_M \).

Comparison of Gradients

Identical diastolic periods were used to compare the gradients. Using a planimeter the mean diastolic gradients (\( \bar{\Delta P}_M \) and \( \bar{\Delta P}_U \)) were determined in each patient. The paired t test was applied to the two sets of values, and the linear correlation coefficient and the mean value of \( \Delta P_M - \Delta P_U \) were determined.

Results

The region of the optimum probe position was identified in less than 60 seconds scan time and was found to be located in the fifth, sixth or seventh intercostal space near the mid-clavicular line. Another 10–15 seconds were required to determine the optimum position. In reviewing the audio signals of the recorded frequency shifts, the beat-to-beat quality seemed uniform in any patient, with only occasional cardiac cycles marred by excessive background noise. This noise probably resulted from slight movements of the
probe on the patient’s skin. In the hard copies of the frequency analyses, the diastolic time course of $\Delta f_{\text{MAX}}$ was relatively well-defined in all patients. However, this time course had to be drawn by hand as a line that enveloped the frequency shifts due to the diastolic blood velocities; there was some uncertainty in choosing the location of this line. From the frequency analyses we estimated that this uncertainty was $\pm 0.125$ kHz (SD); according to equation 4 this corresponds to $\pm 0.4$ mm Hg (SD) and $\pm 0.6$ mm Hg (SD) at gradients of 5 and 10 mm Hg, respectively. In the patients with aortic or mitral valve insufficiency (table 1) no unusual features were noted in the audio signal or in the hard copies of the frequency analyses.

In all patients $\Delta f_{\text{MAX}}$ increased rapidly in early diastole (accelerative flow phase) and reached a peak
value within 0.15 seconds after the onset of diastolic flow (figs. 1–4, panel B). In the patients in atrial fibrillation $\Delta f_{\text{MAX}}$ then decreased relatively smoothly (decelerative flow phase) toward its end-diastolic value. In the two patients in sinus rhythm $\Delta f_{\text{MAX}}$ reached a second peak value in late diastole. The opening and closing motions of the disc were prominently represented in the frequency analyses as blackened vertical columns that flanked the diastolic frequency shifts.

In most patients the left ventricular pressure tracings exhibited early diastolic oscillations (figs. 1–4, panel A); in some, the oscillations persisted throughout diastole. In one patient (patient 9, table 1) the left ventricular pressure tracing was smoothed to the extent that excessive damping was evident. The left atrial pressure tracings did not exhibit similar oscillations. In one patient (no. 6) excessive damping was evident.

In the constructed gradients (figs. 1–4, panel C) the effects of the oscillations in the left ventricular pressure tracings were prominent in the time course of $\Delta P_M$, while $\Delta P_U$ was devoid of similar features. Early diastolic peak $\Delta P_M$ exceeded early diastolic peak $\Delta P_U$ in all patients except patient 9; the mean difference...
between the two peak values was 6.5 ± 6.8 mm Hg (SD). In the nine patients \( \Delta P_M \) ranged from 2–12.5 mm Hg and cardiac output from 3.1–7.5 l/min (table 1). The linear correlation coefficient between \( \Delta P_M \) and \( \Delta P_U \) was 0.96 and \( t = 1.23 \). The mean value of \( \Delta P_M - \Delta P_U \) was 0.3 ± 1.0 mm Hg (SD).

Discussion

Torricelli’s law (equation 1) is a steady-flow orifice equation; thus, \( \Delta P_U \) (equation 4) ignores the inertial component of the pressure gradient and will, at best, only predict the dissipative (frictional) component. A good agreement between the instantaneous values of \( \Delta P_M \) and \( \Delta P_U \) can therefore not be expected. Theoretically, the pressure gradient would be expected to exceed the dissipative component during the accelerative flow phase, while the reverse would be expected during the decelerative phase. The finding that early diastolic peak \( \Delta P_M \) exceeded early diastolic peak \( \Delta P_U \) is probably due in part to the neglect of the inertial component. When the entire diastolic period is considered, inertial effects will theoretically tend to cancel each other, and this phenomenon is probably reflected in the finding that the mean diastolic gradients agreed closely. This latter finding indicates that Torricelli’s law predicts the dissipative component of the gradient closely in the Björk-Shiley mitral implant.

Using the cosine function, deviations in the probe position from the ideal position can theoretically result in a significant underestimation of the actual gradient. However, if the deviation is less than 15–20°, the errors will be relatively small, because the values of the cosine function remain close to unity in this range (\( \cos 15^\circ = 0.966, \cos 20^\circ = 0.939 \)). One of the primary goals of our investigation was to determine whether the disc structure in the implant directs the mitral flow posteriorly and thus precludes a probe position compatible with \( \cos \theta = 1 \). The results do not, however, demonstrate any significant tendency for \( \Delta P_U \) to underestimate \( \Delta P_M \), so it appears that the presence of the disc does not prevent the attainment of a satisfactory probe position. Bone, cartilage and lung tissue will absorb the incident sound beam, and therefore only a limited number of probe positions on the precordium will allow acoustic contact with the flow through the valve. It is unlikely that a satisfactory probe position could have been attained in the majority of the patients if the velocity vectors of \( V_{\text{MAX}} \) all pointed in the same direction. The results indicate that there are a number of jets issuing from the valve in

![Figure 4. Data and results from patient 2 (sinus rhythm). A) Pressure tracings. B) Frequency analysis. C) Constructed gradients. Symbols as in figure 1. Note effects of atrial contraction.](http://circ.ahajournals.org/doi/10.1161/01.CIR.80.1.441)
The manometric registrations are also subject to error. In early diastole the left atrial pressure falls rapidly, and if the pressure registrations are damped, it will result in an overestimation of the actual gradient. Similarly, a damped left ventricular pressure tracing may result in underestimation of the actual gradient. The damped pressure registrations observed in patients 6 and 9 may be responsible for most of the discrepancies between \( \Delta P_M \) and \( \Delta P_U \) in these two patients. The oscillations observed in the left ventricular pressure tracings are probably the result of artifacts introduced by the mechanical components in the pressure transmission lines of the manometric system. The early diastolic peak value of \( \Delta P_M \) therefore may be significantly larger than the actual peak gradient in many of the patients. The ultrasound system is not subject to similar frequency response difficulties because its signal transmission line is devoid of mechanical components.

The rapid changes in diastolic flow through the valve indicated by the time course of \( \Delta f_{MAX} \) may induce significant pressure gradients within the left atrium and left ventricle. Thus, the concept of a pressure gradient becomes more complex and the gradient registered by the manometric system may depend on the localization of the catheters within the chambers. Possibly, such a situation may account for some of the discrepancies between \( \Delta P_M \) and \( \Delta P_U \) in the present investigation.

The frequency shifts caused by regurgitant mitral blood streams (prosthetic valve leakage and paravalvular fistulae) are rarely detected. The major part of such blood streams are shielded from the incident sound beam by the structures of the valve, so the exposed volume of high velocity blood is apparently too small to allow registration by the ultrasound system used in our investigation.

The frequency shifts from the regurgitant blood stream in aortic insufficiency are frequently detected from probe positions on the precordium. These frequency shifts have low signal strength, are disorderly (noise-like), and \( \Delta f_{MAX} \) is poorly defined. The regurgitant stream in aortic insufficiency might be confused with the mitral blood stream, but in our experience this has not been a problem.

Our results indicate that the ultrasound technique determines the mean diastolic pressure gradient in the Björk-Shiley mitral implant with constancy and an accuracy comparable with manometric methods. Errors were evident in the manometric registrations, and the technique is probably more accurate than indicated by the comparison between \( \Delta P_M \) and \( \Delta P_U \). We estimated that the value of \( \Delta P_u \) could be ascertained with an accuracy of approximately \( \pm 0.5 \) mm Hg (SD). The actual mean diastolic gradient may be determined with an equal accuracy.

\( \Delta P_U \) ignores the inertial component of the gradient and the instantaneous value of \( \Delta P_U \) will therefore deviate from the actual gradient whenever rapid changes in blood velocity occur. The largest deviations may be expected during the accelerative flow phase in early diastole.

References

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