SKIN-COUPLED PVDF MICROPHONES FOR NONINVASIVE VASCULAR BLOOD SOUND MONITORING

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Abstract—Vascular access is the "Achilles Heel" of hemodialysis, as maintaining high flow characteristics (access patency) is critical to achieving efficient dialysis treatment. Thus, monitoring of vascular access is essential for maintaining long-term dialysis success. Blood sounds change in the presence of stenosis and can be analyzed digitally as phonoangiograms (PAGs) to determine changes in hemodynamic flow. We propose a multi-channel PAG recording sensor suitable for rapid, non-invasive vascular access monitoring. Here we present the initial design and characterization of sensors appropriate for recording PAGs from the skin surface. An optimized sensor size and backing material was selected to improve sensitivity and to provide a neutral frequency response. The sensor performance was finally compared with a conventional stethoscope on a controlled blood flow stenosis benchtop phantom.

I. Introduction

Access dysfunction accounts for 20-30% of hospital visits for dialysis patients and the loss of access patency can double their mortality risk [1]. The predominant causes of access dysfunction are stenosis (vascular narrowing) and thrombosis (vascular occlusion), which have a combined incidence of 66-73% in Arteriovenous fistula (AVFs) and 85% in Arteriovenous Grafts (AVGs). Thus, monitoring access function at the point of dialysis may be able to reduce emergency interventions [2]. Prior studies have found a robust link between PAG temporal-spectral variation and degree of stenosis (DOS) [3-5].

A simple, reliable, and automated dialysis access monitoring tool which could be placed on a patient's access prior to dialysis would provide real-time tracking of access health, but does not exist [6]. The current standard of care includes using a stethoscope for physical examination of an access, but existing models are not suitable for surveillance over large vascular accesses. We propose that an array of skin-coupled sensors can be used to measure PAGs at the point of dialysis care, and that the acoustic properties of PAGs can be analyzed to estimate DOS (Figure 1).

Polyvinylidene fluoride (PVDF) is a piezoelectric polymer that is mechanically flexible and sensitive to applied strain [7]. It has been used to realize contact microphones for monitoring of cardiorespiratory information such as respiration and heartbeat [7]. Here we explore PVDF's capacity to record PAGs and determine appropriate sensor dimensions and backing materials in order to improve the signal to noise and distortion ratio (SNDR) and frequency response of the sensor.

II. Sensor Construction

Sensors were constructed using 28-µm silver ink metallized PVDF film. Films were sandwiched between two identical printed circuit boards with annular electrical contacts surrounding drilled holes to electrically contact each side of the film separately. Circuit board electrodes were attached to the PVDF film using silver conductive epoxy adhesive (MG Chemicals, Model 8331). The skin-facing side of the PVDF was electrically grounded and covered in a thin PDMS film (Ecoflex 00-10) with a similar mechanical impedance to muscle. The opposite side of the PVDF film used one of three backing materials to constrain the mechanical resonant modes and control the acoustic response. The three backings included: a PDMS layer (Ecoflex 00-10), silicone gel (Dow Corning SYLGARD 527 dielectric gel) plus polyimide tape, or an open backing (air-backed). Four sensor diameters were also produced—2, 4, 8, and 16 mm—to study the effect of sensor size on acoustic sensitivity and frequency response.

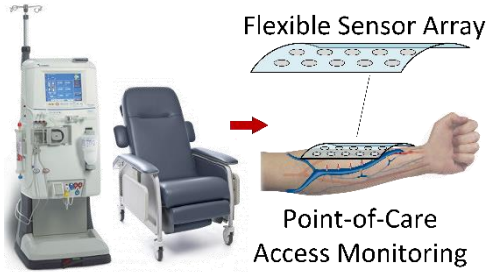

Figure 1. Dialysis vascular access monitoring with a flexible array of sensors capable of recording and processing blood flow sounds to determine stenosis risk.

III. Benchtop Characterization

All sensors were tested under the same conditions to determine the best combination of sensor size and backing material, and their responses were compared to a conventional stethoscope. Characterization tests

included slow frequency sweeps to determine the acoustic frequency response and single-tone tests to determine SNDR. Functional tests involved signal recordings from a vascular flow phantom to simulate the recording of hemoacoustics. Sensor data was collected using LabVIEW at sampling rate of 10 kHz. Sensor data was compared to data collected from a digital recording stethoscope (3M Littmann Stethoscope 3200). The sampling rate of the stethoscope was 4 kHz, but this was digitally upsampled to 10 kHz to match the sample rate of the PVDF sensors for frequency analysis.

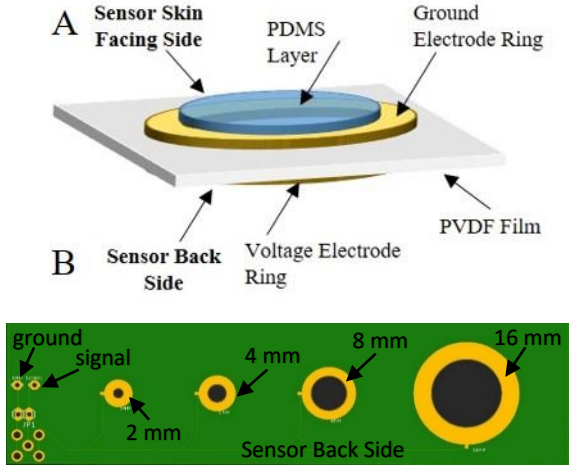

Figure 2. (A) Schematic structure of a sensor showing the PVDF film sandwiched between two electrodes and a layer of PDMS on the skin-facing side. (B) Layout of a sensor test board containing 2, 4, 8 and 16-mm diaphragm sizes.

A. Frequency Response

A frequency generator was used to generate a linear frequency sweep from 20 Hz to 5 kHz over 60 sec. The output was connected to a contact speaker element to generate acoustic vibrations through a 6 mm layer of PDMS rubber. This arrangement mimicked the typical thickness of tissue over a blood vessel in a vascular access. To account for variations in surface coupling pressure and gain differences, recordings from sensors and the stethoscope were normalized to a −10 dB RMS level in MATLAB. Power spectral densities from this test revealed the frequency responses of the PVDF sensors and the stethoscope. Generally, PVDF sensors had a flatter frequency response and wider bandwidth.

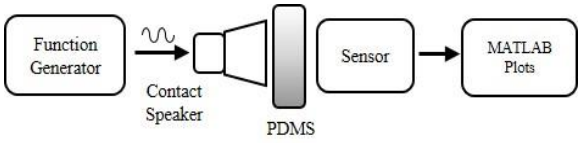

Figure 3. Test setup for frequency response measurements.

To compare the relative frequency response of PVDF sensors to the stethoscope, the power spectral density of the latter was subtracted from each sensor response (Figure 4) to normalize the sensor response to that of the stethoscope. From this analysis several effects were evident. First, the PVDF sensors had relatively less low-frequency response when compared to the stethoscope, with an average 0-dB crossover at approximately 30 Hz. Second, they had significantly greater sensitivity in the range of 30 – 300 Hz, and also above 700 Hz. Finally, of the PVDF sensors, those with silicone gel backing had an additional 3 dB gain in most frequency ranges.

B. Single-tone response

Single tone testing was done at 3 frequencies (150, 300, and 450 Hz) using the same setup (Figure 3). To compare the response between sensors and the stethoscope, the SNDR was calculated (Table 1). On average, all PVDF sensors outperformed the stethoscope, with the 2-mm sensor plus silicone gel backing showing the best overall SNDR (Figure 5).

C. PVDF and Stethoscope Recording of Hemoacoustics

To simulate how the sensor would perform on a real patient (but in a controlled environment), recordings

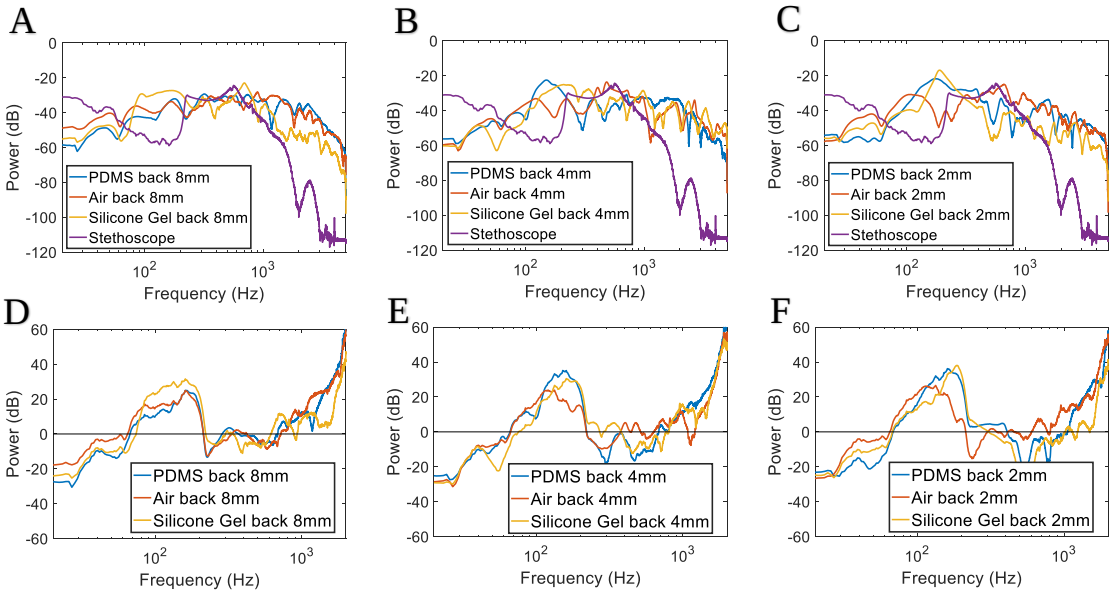

Figure 4. Frequency response comparison between sensors with different backing layers (A-C). When plotted relative to the stethoscope response, PVDF sensors showed enhanced mid-band and high frequency gain (D-F).

were made on a vascular access phantom. The phantom consisted of a 6-mm silicone tube embedded in PDMS (Ecoflex 00-10) at a 6-mm depth. Stenosis was simulated in the center of the phantom with a band tied around the tube to produce an abrupt narrowing.

The phantom was connected to a pulsatile flow pump to simulate human hemodynamics at 432 ml/min and 1120 ml/min (low and high flow rates, respectively). The 2-mm PVDF sensor with silicone gel backing showed similar signal recording quality to the stethoscope (Figure 7) till 100 Hz. At higher frequencies the stethoscope was relatively less sensitive.

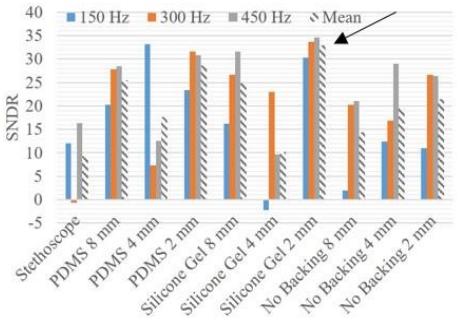

Figure 5. Sensors with 2 mm diaphragms had the highest SNDR and performed well at all three test frequencies.

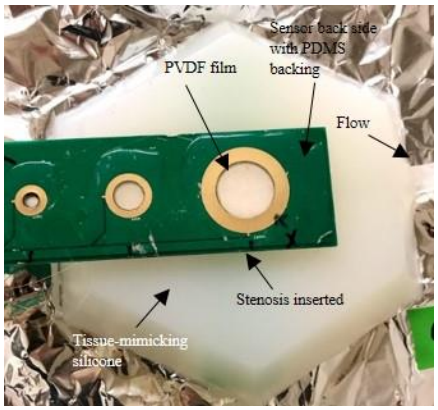

Figure 6. Banded silicone tubing embedded in silicone rubber was used as a vascular access phantom. Water was pumped through the phantom at physiologic flow rates and PAGs were recorded using each type of sensor.

D. Autoregressive modeling of PAG systoles

PAG signals can be processed in multiple ways to enhance the systolic pulse waveform. Additionally, many auditory features can be extracted from the PAG, including the auditory spectral centroid (ASC), auditory spectral flux (ASF), and modeled frequency spectrum using linear predictive coding.

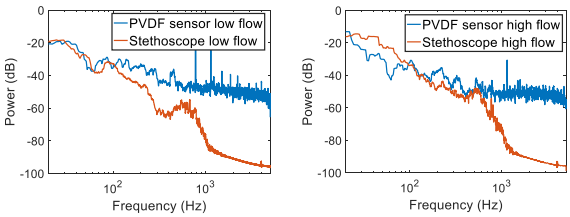

Figure 7. The frequency spectra of PAGs recorded using i) the 2 mm sensor with silicone and polyimide tape backing, and ii) the stethoscope, at (a) low, and (b) high flow rates.

The systole portion of flow can be enhanced automatically using a customized filter based on sub-band frequency domain linear prediction (FDLP) [2]. When used on a broadband transform, FDLP models the Hilbert envelope of a signal. By controlling the number of poles used in the predictive model, significant smoothing of the modeled envelope is obtained. The resulting envelope further approximates the ASF calculated as the L^2 norm of the wavelet transform,

$$ASF[n] = \frac{1}{K}\sqrt{\sum_{k=1}^{K}(|\boldsymbol{W}[k,n]| - |\boldsymbol{W}[k,n-1]|)^2}, \qquad (1)$$

Where $\boldsymbol{W}[k,n]$ is the continuous wavelet transform obtained over K total scales of the PAG.

Similarly, FDLP envelopes are calculated from the discrete-cosine transform (DCT) values of the PAG, $X_{DCT}[k,n]$ over K frequency coefficients. The DCT approximates the envelope of the Discrete Fourier Transform. This implies that the spectrogram of the DCT (treating the DCT as a time sequence) mirrors the time-domain spectrogram around the time/frequency axes.

Just like time-domain linear predictive coding models the frequency-domain envelope, FDLP estimates the temporal envelope when applied in the frequency domain. This implementation uses linear predictive coding (LPC) to model the spectral envelope using a p^{th}-order, all-pole FIR filter

$$\hat{x}[n] = \sum_{k=1}^{p} a_k x[n-k], \text{ i.e. } P(z) = \sum_{k=1}^{p} a_k z^{-k}. \qquad (2)$$

where p is the order of the filter polynomial and $P(z)$ is its z-transform. LPC uses least-squares iterative fitting to determine the coefficients a_k of the FIR filter $P(z)$ such that the error in determining the next value of a series $\hat{x}[n]$ is minimized. The calculated filter is an autoregressive model with significantly lower variance

TABLE 1. MEASURED SNDR OF PVDF SENSORS AND STETHOSCOPE AT 150, 300, AND 450 HZ

	Stethoscope	PDMS Backing			Silicone Gel Backing			No Backing		
		8 mm	**4 mm**	**2 mm**	**8mm**	**4 mm**	**2 mm**	**8 mm**	**4 mm**	**2 mm**
150 Hz	11.9937	20.3000	33.2125	23.3171	16.1953	−2.1913	30.2868	1.9916	12.4593	10.9727
300 Hz	−0.6459	27.7621	7.2581	31.5992	26.6204	23.0329	33.7264	20.2204	16.7918	26.6825
450 Hz	16.3006	28.5185	12.5628	30.8861	31.5639	9.6512	34.5464	21.0196	28.9440	26.4480

than the Hilbert envelope or computed ASF when used for PAG analysis (Figure 8)

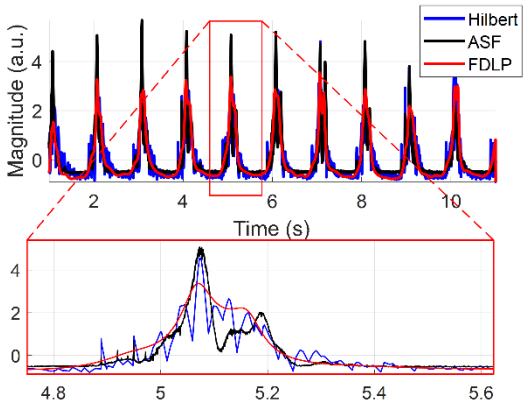

Figure 8. Analytical signals obtained from PAGs can include Hilbert envelope, auditory spectral flux, and FDLP-modeled envelopes. Of these signals, FDLP produced the smoothest response, permitting efficient feature extraction.

When applied to PAGs, FDLP can be used for systolic pulse enhancement [2], or to produce an analytic signal for feature extraction, e.g. to estimate flow variations. Here, we apply the FDLP envelope as a systole enhancement filter by multiplying the envelope by the original PAG signal to apply time-based signal shaping.

Consider recordings from the 2-mm, silicone gel sensor processed using the FDLP systole enhancement compared to a conventional stethoscope recording from the same phantom processed with the same FDLP systole enhancement filter (Figure 9). Spectrograms were computed over 6 octaves with 12 voices/octave, starting at scale 3. The combination of the improved frequency response of the 2-mm sensor and the FDLP processing enhances the systole dramatically and reduces inter-systole noise.

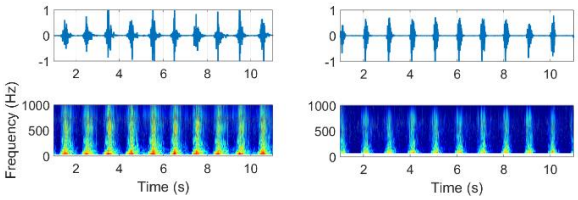

Figure 9. Time and wavelet scale representations of PAG recorded from (A) conventional stethoscope, and (B) the 2 mm sensor with silicone gel, both using FDLP systole pulse enhancement.

IV. Discussion

Our results show that PVDF sensors acoustic response is approximately 10 dB/decade lower than stethoscope acoustic response. However, in the 150–450 Hz range, single-tone testing revealed that PVDF sensors had generally better SNDR. The former also have a wide and relatively flat frequency response to 5 kHz, although the stethoscope acoustic response is likely limited by built-in signal processing to reduce noise pickup. Overall, we conclude that a 2-mm PVDF diaphragm with silicone gel backing forms a reliable transducer for skin-coupled recording of PAGs. In addition, the proposed construction method can be easily modified for use with flexible, polyimide printed circuits to enable flexible arrays of skin-contact microphones. These array microphones could enable point-of-care monitoring of vascular access and could leverage multi-channel signal processing for rejection of interference and new PAG analysis features.

V. Conclusion

We developed a sensor using PVDF film to record blood flow sounds as PAGs that can be used to monitor the functioning of the vascular access. The geometry of the sensor diaphragm and backing options were compared to PAGs recorded from a commercial digital stethoscope. Results from benchtop testing suggest that a 2 mm diameter sensor backed with silicone gel performs favorably when compared to conventional stethoscopes. Future work will include developing a flexible sensor array to enable multi-site recordings to calculate stenosis location, degree, and hemodynamic flow velocity.

Acknowledgments

This work was supported in part by RX001968-01 from US Dept. of Veterans Affairs Rehabilitation Research and Development Service. The contents do not represent the views of the US Government.

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