Detrusor Pressure Estimation from Single Channel Bladder Pressure Recordings

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Abstract—Cystometry measures the behavior of the bladder and is frequently used to evaluate lower urinary tract abnormalities. Cystometry is conducted using two catheters, one in the bladder and the other in the vagina or rectum, which increases discomfort and complexity of the test. In this work we evaluated a method to estimate detrusor pressure (PDET), the pressure generated by a bladder contraction, from only a single catheter measuring vesical pressure (PVES). Using twenty urodynamic studies, we used statistical inference and wavelet multiresolution analysis to maximize the correlation coefficient (R) between estimated PDET and calculated PDET after detecting and eliminating artifacts. Moreover, the estimator design considered a prospective real-time implementation. Root mean square (RMS) error and correlation coefficient were used to evaluate algorithm accuracy in estimating PDET, while a statistical F-score evaluated the accuracy of artifact detection. The output of the proposed estimator compared with calculated PDET, and overall estimation performance showed that RMS = 10.7 ± 2.1 cmH2O and R = 0.88 ± 0.6 (N=20). Moreover, detection accuracy for cough and Valsalva events were 99.5% and 84.3%, respectively. We conclude that estimating PDET from PVES only is feasible making single channel cystometry a possibility.

Keywords—Event-driven estimation, bladder pressure, discrete wavelet transform.

I. INTRODUCTION

The two functions of the urinary bladder are to store and evacuate urine. Cystometry (a type of urodynamics test) assesses reflex function of the bladder during the storage phase, providing information, e.g. on the absence or presence of detrusor instability and detrusor-sphincter coordination [1, 2]. Cystometry uses a rectal or vaginal catheter and a transurethral bladder catheter to measure abdominal (PABD) and vesical or bladder (PVES) pressures, respectively (Fig. 1). PABD describes the forces exerted by abdominal musculature and the surrounding organs on the bladder. PVES measures the pressure within the bladder, which is a summation of the pressure generated by the detrusor muscle (PDET) superimposed on PABD. The detrusor pressure is then distinguished from abdominal induced changes by the simultaneous difference between vesical and abdominal pressure through the formula [2]:

\[ P_{DET} = P_{VES} - P_{ABD} \]  

(1)

In clinical practice, cystometry is restricted to a short length of time due to patient comfort and facility time. Ambulatory recordings of bladder pressure (ambulatory urodynamics) have also been demonstrated to provide new physiological insights. For example, Damaser et al. showed that the difference between full bladder pressure and almost empty bladder pressure can be used to estimate the detrusor pressure of a limited set of patients who undergo clean intermittent catheterization [3].

One challenge in urodynamic recordings (both stationary and ambulatory) is abdominal pressure artifacts due to the movement of the abdominal reference catheter. Only if the abdominal catheter perfectly measures PABD and PABD is exactly transmitted to PVES is the PDET estimated from (1) valid; artifacts in PABD instantly corrupt PDET. Urodynamics therefore requires skill, patience, and experience to gather and interpret bladder pressure recordings.

As (1) indicates, a single measurement of PVES contains both the signal of interest (PDET) and the PABD signal. The goal of single-channel urodynamics is to measure only PVES, then use signal processing to extract PDET after removing the PABD signal. Besides the cost savings of a single catheter used for the test, this technique would increase comfort, reduce infection risk, and reduce the complexity of the clinical cystometry exam.

In this paper, we introduce a technique using discrete wavelet multiresolution analysis and statistical inference...
analysis to estimate the detrusor pressure. The paper introduces the proposed framework, presents the signal processing methodologies, and discusses results from an initial analysis of pre-recorded urodynamic signals.

II. PROPOSED FRAMEWORK

Because $P_{DET}$ is generated by detrusor smooth muscle, there are physiological limitations on the speed and force of muscle contraction. In other words, abdominal pressure data and detrusor events may be spectrally distinguished. Previous work in bladder event identification on single-channel recordings showed that wavelet-based algorithms can identify urologically meaningful events solely using $P_{VES}$ with minimal false positives [5, 6].

However, prior work focused on event detection, not $P_{DET}$ reconstruction [5, 6]. Estimating the underlying $P_{DET}$ from a signal containing superimposed $P_{ABD}$, with few distinguishing frequency components, is challenging. Broadly, $P_{VES}$ may be split into $P_{DET}$ and $P_{ABD}$ components with high- and low-pass filtering. Karam et al. demonstrated that by applying a discrete wavelet transformation (DWT) with Daubechies 4-tap wavelet, the signal may be separated into approximation and detail coefficients, where the detail coefficients tend to contain $P_{ABD}$, and the approximation coefficients tend to contain $P_{DET}$ and may be used to detect voiding events [6].

Correlation analysis from 20 urodynamic (UDS) recordings demonstrated that $P_{DET}$ and $P_{VES}$ were consistently linear, indicating that $P_{DET}$ is contained within $P_{VES}$ (Fig. 2). As expected, the correlation of $P_{DET}$ with $P_{ABD}$ was low, indicating that $P_{ABD}$ is independent of $P_{DET}$ and should be removed from the estimate of $P_{DET}$. The proposed framework therefore uses $P_{VES}$ to estimate $P_{DET}$.

The framework comprises three main stages: the prefilter stage, discrete multiresolution analyzer, and signal reconstruction (Fig. 3). The framework is intuitive from the perspective of an embedded system which would ease verification and troubleshooting in an embedded system implementation where $P_{VES}$ is buffered, then pushed to the pipeline using the sliding window/frame technique. The function of each stage is affected by the window size. Therefore, the size of the sliding window ($W_x$) is crucial for estimation efficiency. Selecting the optimal size of the sliding window is outside the scope of this work and was considered as a fixed size ($W_x=32$ sample).

III. METHODOLOGY

A. URODYNAMIC DATA

A total of 20 (7 male and 13 female) cystometry UDS recordings were collected using air-charged catheters with a CT3000Plus Complete Urodynamics system (SRS Medical, N. Billerica, MA) at a sampling rate of 10 Hz. Recordings consisted of multiple signals including volume voided (Volume), intravesical pressure ($P_{VES}$), intra-abdominal pressure as measured via a rectal catheter ($P_{ABD}$). A signal representing the simultaneous difference between these two pressure channels ($P_{DET} = P_{VES} - P_{ABD}$) was calculated. Data were manually annotated to denote relevant events such as cough, valsalva, position change, and voiding contraction during retrograde bladder filling and voiding.

B. PREFILTERING STAGE

The prefilter removes white noise and spike artifacts which arise from movement of catheters during cystometry. A two-stage filter was used. The first stage was a $10^6$ order finite impulse response band-pass filter with bandwidth of 0.02 - 5 Hz and stop band attenuation of 30 dB. The second stage was a 3rd order Savitzky-Golay smoother which used convolutional linear least squares to fit successive adjacent data points with a low degree polynomial [7, 8]. Unlike simpler smoothing techniques, such as a moving average filter, the Savitzky-Golay smoother preserves data features such as peak width and height.

C. MULTiresOLUTION WAVELET ANALYSIS

Figure 2. Correlation between $P_{ABD}$ and $P_{DET}$ (left) as well as $P_{VES}$ and $P_{DET}$ (right) from data from 20 human subjects.
The filtered data were passed to a wavelet multiresolution analyzer (MRA) and decomposed into fundamental frequency components. The strength of MRA dwells in its ability to preserve time-domain and frequency-domain information using the DWT. The proposed MRA unit was augmented with a tunable weight vector \( W_i \), where \( i \in [1, 2, \ldots, N] \) and \( N = \log_2(W_{sz}) \) was the wavelet resolution level. The weight vector provided a flexible method of estimating \( P_{DET} \) using a weighted sum. In this work, we used \( W_{sz}=32 \). Since the sampling rate of UDS datasets was 10 Hz, it was possible to decompose the 32-sample frame into five frames representing the window component at 1, 2, 3, 5, and 10 Hz (32 sample in each resolution).

In addition to window size, sampling frequency, and weighting vector, selecting a mother wavelet for MRA played an important role in estimating \( P_{DET} \). The output of the wavelet decomposer, \( \hat{P} \), can be described as:

\[
\hat{P}(W, \psi, W_{sz}) = \sum_{i=1}^{N} \left( \frac{1}{\sqrt{2}} \right)^i W_i \psi_i
\]

where \( W_i \) is the window resolution at level \( i \), \( \psi(.) \) is the mother wavelet function, and \( N = \log_2(W_{sz}) \in \mathbb{R} \) is the depth of decomposition. Usually, the vector \( W_{sz} \) is predefined and selected to capture the slowest event (e.g., Valsalva). Instead, we investigated varying the mother wavelet first with a unity weighted vector value. Then the \( W \) value was obtained with a static mother wavelet function via least mean square (LMS) fitting.

Reconstructing a 10 min cystometry signal from its MRA resolutions provided a test for wavelet function fidelity. The RMS error between the original and reconstructed signal called reconstruction error (RE) and the reconstruction computation time (RT) were used to evaluate the selected wavelet function though the reconstruction test. A set of popular symmetrical and orthogonal wavelet functions were selected to test as shown in Table 1. The Symlet wave function with four vanishing points (Sym4) was selected since it achieved the best balance between RT and RE.

In order to find the weight vector, LMS was used with \( P_{VES} \) as the input and \( P_{DET} \) as the output. With \( W_{sz}=32 \) and wavelet function Sym4, the \( P_{VES} \) was decomposed into 5 levels. The LMS results showed that the last level weight was significantly higher than the other weights when estimating the low-frequency \( P_{DET} \) signal (Fig. 4).

### 3.4 EVENT DETECTION AND ELIMINATION

Event detection improves the DWT estimate of \( P_{DET} \) because it allows an event-dependent selection of the reconstruction weights \( W_i \). For example, if an abdominal event (containing high frequency elements) is detected, the weighting vector for windows during the event can de-emphasize initial scales of the DWT.

![Figure 4. Gains in wavelet decomposition of 5-level wavelets resulting from LMS for the cystometry dataset and 32 sample window size.](image)

Statistical features of the window serve as an alternative abstract representation of the window and a strong indicator of contraction onset and termination of each event. Previous work reported using an adaptive threshold that tracks the baseline pressure and detects when the \( P_{DET} \) estimate exceeds this plus a fixed threshold [9]. The most recent work has investigated the detection and classification of bladder events using wavelet analysis [5, 6].

However, while previous studies reported only detection of event onset, detecting the total span of an event is an important factor that has received less attention [7]. The span of the event is the time difference between the onset and termination time. The given UDS dataset showed that the average duration of coughs was 1.4±0.5 sec and the average duration of a Valsalva event was 2.5±0.5 sec. The fixed window size of 32 samples was sufficient to capture the slowest event with at most two consecutive windows.

Event detection used the previous observations to identify a vector of threshold values to identify the event onset and termination. For a sliding window, the kernel of the event detection algorithm was:

\[
\Gamma(k) = \begin{cases} 
F_y > T_m \text{ and } f(v) > T_r \\
0 & \text{otherwise}
\end{cases}
\]

where \( v = [\mu, \sigma, \xi, \delta] \in \mathbb{R}^4 \) is a vector, \( \sigma \) is the local window standard deviation, \( \mu \) is the arithmetic mean, \( \delta \) is the maximum value of local gradient, \( \xi \) is the rate of signal zero crossing, \( F_y \) is a flag set at onset and cleared at termination, and \( T_m \) and \( T_r \) are threshold values corresponding to the average duration of each event. In the presented work, \( T_m \) and \( T_r \) were empirically determined by manually identifying the lengths of valsalva and cough events.

| Table 1. The characteristic of reconstruction outcomes using different wavelet functions |
|---------------------|------|------|--------|-------|-------|
|                   | Haar | DB4  | Sym4  | Dmey | Coif4 |
| RT (msec)         | 1.9  | 3.7  | 3.4   | 8.7   | 14    |
| RE                | 1e-5 | 1e-12| 1e-12 | 1e-12 | 1e-12 |

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In general, the estimated $P_{DET}$ was calculated by eliminating the detected events, i.e:

$$P_{DET}(W, W_{sz}, \mu, \sigma, \delta, \xi) = \begin{cases} \frac{p}{p - \mu} & \Gamma(k) = 1 \\ p - \mu & \text{otherwise} \end{cases}$$

(4)

This relatively simple technique was adopted due to the low computational overhead, which makes it more suitable for real-time implementation.

IV. RESULTS AND CONCLUSION

In addition to smoothing the $P_{VES}$ signal, the 2-stage prefilter attenuated large amplitude, pressure spikes due to coughs (Fig. 5). Coughs should only appear in $P_{ABD}$, with no pressure in the $P_{DET}$ estimation, because they are generated entirely by the abdominal muscles. Pairing the wavelet decomposer with the prefilter resulted in a 23% reduction in cough artifacts passing to the $P_{DET}$ estimation instead of $P_{ABD}$.

The root main square ($RMS$) value and correlation coefficient $R$ were used to evaluate algorithm accuracy. Furthermore, a statistical F-Score was used to evaluate the accuracy of artifact detection. The output of the estimator was compared with the calculated $P_{DET}$, and the overall estimation performance showed $RMS = 10.7 \pm 2.1 \text{cmH}_2\text{O}$ and $R = 0.88 \pm 0.6$ (N=20). Moreover, detection accuracy for cough and valsalva were 99.5% and 84.3% respectively (Fig. 6). Detection accuracy was determined relative to clinical annotation of events for each cystometry recording.
In conclusion, we have developed a promising proof-of-concept of estimating $P_{\text{DET}}$ using data from a single catheter. The single catheter estimator could mitigate the burden of using two catheters during cystometry.

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