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Myometrium electromechanical modeling for internal uterine pressure estimation by electrohysterography

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***Abstract*— During delivery, quantitative information on the uterine activity can be provided by internal uterine pressure (IUP) recordings using an invasive intrauterine pressure catheter (IUPC). The electrohysterogram, which measures the electrical signal that drives the mechanical contraction of the uterine muscle and the consequent IUP increase, is recorded by electrodes placed on the abdomen. Recent work demonstrated the possibility of reliably estimating the IUP noninvasively by electrohysterographic (EHG) signal analysis. To further improve the accuracy of IUP estimates, we investigated the use of three nonlinear functions for modeling the relationship between the electrical activation measured by the EHG signal and the mechanical response of the uterine muscle. The feature employed for obtaining a first estimate of the IUP is the unnormalized first statistical moment of the EHG spectrum. The relationship between the extracted feature and the IUP is modeled by a second-order polynomial, a logarithmic, and an exponential function. For validation, the IUPC and the EHG signals were recorded on nine women in labor. A second-order polynomial model already provided estimates that are highly correlated with the IUPC signal ($r = 0.73$). However, the logarithmic model resulted to be the most accurate, especially in terms of root mean squared error ($\mathrm{RMSE} = 5.13$ mmHg).**

I. INTRODUCTION

During parturition, effective uterine contractions induce progressive cervical dilatation and produce an internal uterine pressure (IUP) increase acting towards the fetus expulsion at the end of delivery. Monitoring the mechanical activity of the uterus provides important information on the efficiency of the contractions. However, existing methods employed in clinical practice impose a compromise between reliability and invasiveness.

Quantitative information concerning uterine functionality can be provided by measuring the amniotic IUP with an internal uterine pressure catheter (IUPC). However, IUPCs are highly invasive and may lead to complications [1]. External tocography, which is an indirect measure of the IUP increase, is noninvasive, but it does not provide accurate information on amplitude and duration of the contractions [1]. A promising noninvasive technique for uterine contraction monitoring is electrohysterography.

The electrohysterographic (EHG) signal, which can be measured by contact electrodes on the abdomen, is associated with the electrical activity that triggers the mechanical contraction of the uterine muscle, the myometrium. For noninvasive estimation of the IUP by EHG signal analysis, optimal linear filtering [2] and root mean squared value analysis [3] have been previously proposed. These methods were recently compared in [4] to a novel EHG-based technique which comprises the derivation of a first IUP estimate as the unnormalized first statistical moment of the EHG signal frequency spectrum. The estimate accuracy is then improved by identifying a second-order polynomial model describing the nonlinear relationship between the electrical activation, which is represented by the unnormalized EHG first spectral moment, and the mechanical contraction of the myometrium, which is measured in terms of IUP.

The second-order polynomial model described in [4], already provides estimates that are highly correlated with the recorded IUP. However, other nonlinear models might be more suitable for characterization of the underlying physiological process.

In the present study, similarly to [4], the feature for a first estimation of the IUP was obtained by the unnnormalized first statistical moment of the EHG signal frequency spectrum. Then, in order to further improve the estimate accuracy, the nonlinear relationship between the electrical activation and the mechanical contraction of the uterus was modeled by three different functions: in addition to a second-order polynomial model, which was used for comparison, we tested a logarithmic function and an exponential function. Simultaneous IUP recordings by an IUPC were employed for quantitative validation of the pressure estimates.

II. METHODOLOGY

Multichannel EHG signals and the IUPC were recorded simultaneously on women during delivery. The first unnormalized statistical moment was calculated in the time-frequency domain from the EHG signal as the feature providing a first IUP estimate. For each model, the model coefficients were then adaptively identified from the invasively recorded IUP separately for each contraction. Eventually, the IUP was estimated by using the median values of the model coefficients. The experimental set-up and the method used for the feature extraction are described in details in [4] and are here only briefly reported.

A. *Experimental set-up*

The experimental data were collected at the Máxima Medical Center, Veldhoven (the Netherlands) from nine women in labor. The length of data recorded from each subject ranged from 22 to 90 min (from 9 to 33 contractions). The EHG was recorded by means of four unipolar contact Ag-AgCl

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electrodes placed on the abdomen below the umbilicus as described in [4]. The common reference of the electrodes was placed on the right hip. In order to obtain an efficient rejection of the electromagnetic interference, a driven-right-leg electrode was placed below the reference, on the right hip, and the cables were actively shielded.

A Koala M1333A (Philips Healthcare, Best, the Netherlands) IUPC was inserted in the uterine cavity to measure the IUP. The IUP and the EHG were simultaneously recorded and digitized at a sampling rate of 1000 Hz with an M-PAQ amplifier (Maastricht Instruments Ltd., the Netherlands). The lower and upper cut-off frequencies of the amplifier were set at 0.01 Hz and 75 Hz, respectively. Due to the low-frequency content of the involved signals [5], the IUP and the EHG signal were down-sampled to 10 Hz after anti-aliasing low-pass filtering. Eventually, the IUP was estimated by five bipolar EHG signals, obtained by subtracting the EHG signals recorded at contiguous electrodes [4].

B. Feature extraction

Due to its non-stationarity, the EHG signal energy can be represented in frequency, f, by the Spectrogram, $\rho(t,f)$, which is obtained by the squared magnitude of the short-time Fourier transform of the signal as a function of the time t, $x(t)$, through a limited time-window $w(t)$, i. e.,

$$\rho(t,f) = \left| \int_{-\infty}^{+\infty} x(\tau) w^*(\tau - t) e^{-j2\pi f \tau} d\tau \right|^2, \tag{1}$$

where $(\cdot)^*$ is the conjugate operator. The force generated by the whole uterus and the consequent IUP increase depend on the level propagation of action potentials from cell to cell and the amount of muscle cells involved in the contraction [6]; in smooth muscle, these two phenomena are mutually dependent, as propagation requires recruitment of multiple cells. A first estimation of the IUP can be therefore derived by the unnormalized first statistical moment $\Psi(t)$ of the Spectrogram in a selected frequency band $[f_{min}, f_{max}]$

$$\Psi(t) = \int_{f_{min}}^{f_{max}} f \, \rho(t,f) \, df. \tag{2}$$

Based on previous studies [1] [4], the selected frequency band was [0.3 Hz, 0.8 Hz].

C. Electromechanical activation modeling

The first unnormalized statistical moment obtained by the Spectrogram can be further processed in order to increase the accuracy of the estimate. In general, the relationship between the uterine electrical activation, the EHG signal, and the mechanical response, the IUP, can be considered nonlinear. In vitro studies on animal uterine strips [7] highlighted, in fact, during induced isometric contractions, a non-linear characteristic of the tension developed by the myometrium as a function of the the electrical stimulation. Furthermore, the IUP increase is the result of the simultaneous contraction of an adequate number of cells and it is associated with a widespread electrical activity of the whole myometrium. Therefore, electrical activities that are local or poorly coordinated can be recorded without being necessarily followed by a linearly related IUP increase.

Previous studies demonstrated that the use of a second-order polynomial model is a suitable representation of the relationship between the unnormalized first EHG spectral moment and the recorded IUPC signal [4]. We indicate by $\Psi_k(t)$ the unnormalized first EHG spectral moment calculated in each channel k and by $\mathrm{IUP}(t)$ the pressure measured by the catheter. An estimate, $\widehat{\mathrm{IUP}}_k(t)$, of the pressure recorded by the catheter can be derived in each channel by a polynomial expansion of $\Psi(t)_k$ with coefficients a_P, b_P, and $c_P \in R$ as given in (3):

$$\widehat{\mathrm{IUP}}_k(t) = a_P + b_P \Psi_k(t) + c_P \Psi_k(t)^2. \tag{3}$$

In (3), a_P models the offset, b_P the gain, and c_P the nonlinearities of the relationship between the estimated IUP in each channel k and the IUPC signal.

The model in (3) is a standard representation of a nonlinear relationship. However, for values of $\Psi_k(t)$ above the maximum point of the parabola described by (3), an IUP decrease is obtained for increasing values of $\Psi_k(t)$; this effect might be not representative of the underlying physiology. Therefore, a logarithmic model, i.e.,

$$\widehat{\mathrm{IUP}}_k(t) = a_L + log(b_L \Psi_k(t) + 1), \tag{4}$$

a_L, b_L, and $c_L \in R$, was considered as a potentially more suitable representation of the electromechanical activation of the myometrium. According to (4), an increase of electrical activation, $\Psi_k(t)$, always produces a increase of the mechanical response, $\mathrm{IUP}(t)$.

In order to account for a possible asymptotic behavior of the IUP in response to increasing electrical activation, an exponential model of the form

$$\widehat{\mathrm{IUP}}_k(t) = a_E + b_E(1 - e^{c_E \Psi_k(t)}), \tag{5}$$

with a_E, b_E, and $c_E \in R$, was also tested.

The coefficients a_P, a_L, and a_E in (3), (4) and (5), respectively, represent the offset between the electrical activation and the IUP. In fact, even in the absence of uterine contraction, the IUPC signal gives a value, referred to as baseline tone, which is usually different from zero. Since the IUPC signal baseline tone is affected by factors that are not related to the uterine activity [8], it was detected by the method described in [3] and digitally removed on the basis of the first two minutes of recording. As the same procedure was applied to $\Psi_k(t)$ prior identification of the model coefficients, the constant terms a_P, a_L, and a_E were set to zero. The other coefficients of the models were identified separately for each channel and each contraction by the Nelder-Mead Simplex search method [9] using the invasively recorded IUP as the reference signal. To this end, contiguous time segment of the IUPC signal, each containing one contraction, were automatically selected using an adaptive threshold [3]. For each channel and model, an estimate of the IUP was eventually derived by using a single set of coefficients, which

TABLE I

AVERAGE RESULTS

Model	$r(p<0.05)$	RMSE (mmHg)
Polynomial	0.73 ± 0.11	13.47 ± 6.67
Logarithmic	0.74 ± 0.11	5.13 ± 4.74
Exponential	0.74 ± 0.10	6.36 ± 5.63

was obtained by the median value of all the coefficients identified separately for each contraction.

D. Evaluation of the estimate quality

The estimates obtained by the three nonlinear models in each channel k, $\widehat{\mathrm{IUP}}_k(t)$, were compared in terms of correlation coefficient r and root mean squared error (RMSE)

$$\mathrm{RMSE} = \sqrt{E(\mathrm{IUP}(t) - \widehat{\mathrm{IUP}}_k(t))^2}, \tag{6}$$

where $E[\cdot]$ is the expected value operator, with the IUPC signal, IUP(t).

For a clinical feasibility assessment, the peak pressure amplitude of each contraction, $\widehat{\mathrm{IUP}}_P$, measured on the IUP estimate was compared to the peak pressure amplitude, IUP_P, derived from the IUPC signal. The peak pressure amplitude, is, in fact, one of the parameters commonly employed in clinical practice for the evaluation of uterine contractility by IUPC recordings [10][11].

E. Results

The average correlation coefficient, r, and the RMSE obtained by the polynomial, the logarithmic, and the exponential model are shown in table I. The results refer to the average and standard deviation calculated over the nine patients and the five bipolar channels.

As from table I, the logarithmic and the exponential model provided significant estimation improvement with respect to the polynomial model in terms of RMSE. Therefore, for clinical feasibility, only the logarithmic and the exponential model were further analyzed.

The mean peak pressure of the IUP estimated by exponential and logarithmic modeling was, respectively, 11.21 ± 5.82 mmHg and 10.4 ± 12.9 mmHg lower that the peak pressure recorded by the catheter. The average IUPC signal peak pressure value was 41.6 mmHg. A more detailed overview of the peak pressure difference between the IUPC signal and the estimates is provided by the Bland-Altman plots in Fig. 1 and Fig. 2, for the logarithmic and for the exponential model, respectively. The IUP peak amplitude difference is shown as percentage of the average between IUP_P and $\widehat{\mathrm{IUP}}_P$.

In Fig. 1 and Fig. 2 all contractions and channels of all subjects were equally considered. If the average value of peak pressure difference was calculated separately for each patient, the mean peak pressure difference resulted in $-9\% \pm 5\%$ and $-18 \pm 9\%$ of the average between IUP_P and $\widehat{\mathrm{IUP}}_P$ for the logarithmic and for the exponential model, respectively.

Fig. 3 and Fig. 4 show an example of the IUPC recording and the corresponding IUP estimated by the logarithmic and

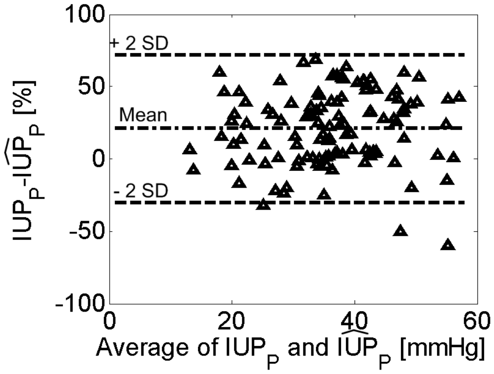

Fig. 1. Bland-Altman plot of the peak pressure, IUP_P, of the IUPC signal, and the peak pressure, $\widehat{\mathrm{IUP}}_P$, of the IUP estimated by logarithmic modeling.

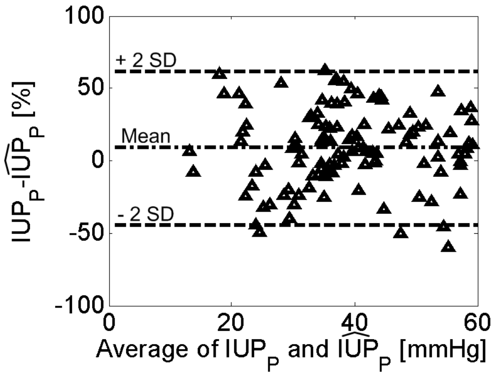

Fig. 2. Bland-Altman plot of the peak pressure, IUP_P, of the IUPC signal, and the peak pressure, $\widehat{\mathrm{IUP}}_P$, of the IUP estimated by exponential modeling.

the exponential model, respectively; the examples refer to the same channel and subject.

III. DISCUSSION

The methods proposed in the literature for noninvasive IUP monitoring, namely optimal linear filtering [2], root mean squared value analysis [3], and the unnormalized first statistical moment of the EHG frequency spectrum in combination with polynomial modeling, have been previously compared to invasive IUPC recordings [4]. According to that study, the use of the first unnormalized statistical model in combination with a second-order polynomial model resulted in the best correlation coefficient ($r = 0.73 \pm 0.11$), but the lowest RMSE (RMSE $= 11.4 \pm 3.12$ mmHg) was obtained by optimal filtering.

The first statistical moment of the EHG signal in the frequency domain is conceived to be representative of the physiology involved in the contraction process; therefore, also in this work, it was derived as the feature providing a first estimate of the mechanical uterine activity. For a more accurate estimate of the IUP, a logarithmic and an exponential model were then compared to the second-order polynomial model already proposed in the literature. The results, obtained from the same data-set as [4], showed that both nonlinear models provide a significant improvement

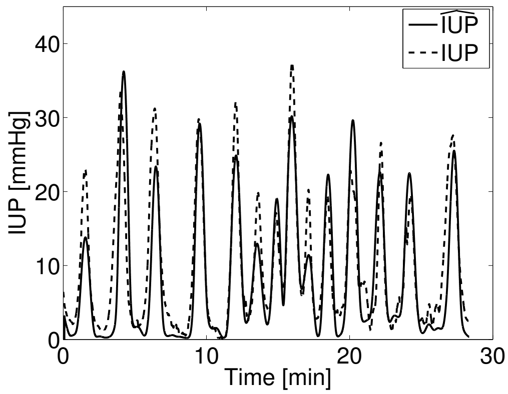

Fig. 3. Example of the IUP estimated by the logarithmic model. For the entire waveform $r = 0.86$ and $\mathrm{RMSE} = 4.5$mmHg.

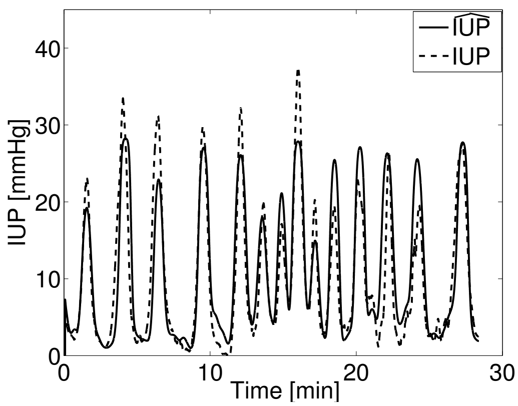

Fig. 4. Example of the IUP estimated by the exponential model. For the entire waveform $r = 0.87$ and $\mathrm{RMSE} = 4.8$mmHg.

with respect of the published methods. When compared to the invasively recorded IUP, the logarithmic and the exponential model provided a higher correlation coefficient than the use of a polynomial model, $r = 0.74 \pm 0.11$ and $r = 0.74 \pm 0.10$, and an average RMSE much lower than optimal filtering, i. e., $\mathrm{RMSE} = 5.13 \pm 4.74$ mmHg and $\mathrm{RMSE} = 6.36 \pm 5.63$ mmHg, respectively.

In the present work, IUPC signals were used for validation because IUPC recordings are currently the most reliable technique for uterine contraction monitoring [1]. Nevertheless, in [12], from the IUP measured by three fluid-filled catheters simultaneously inserted in the same uterus, the authors found a measurement uncertainty up to 25%. In [11], where transducer-tipped catheters were used, a lower measure uncertainty was found; for catheters inserted independently in the same uterus, the mean peak pressure difference (bias) for each patient was about $9.96\% \pm 9.7\%$ of the average value recorded by the two catheters.

In the present study, when comparing peak pressure measurements derived from the estimate of all patients and channels to the invasively recorded IUP, both models resulted in a biased estimate with a standard deviation of about 25%. However, by comparing the invasive catheter recording and the noninvasive IUP estimated by EHG signal analysis in terms of average peak difference per patient, as it was done in [11], we obtained an average peak pressure difference for the logarithmic model of $-9\% \pm 5\%$, which can be comparable to the measurement uncertainty ratios reported for transducer-tipped IUP catheter recordings in [11].

IV. CONCLUSIONS AND FUTURE WORKS

The use of either a logarithmic or an exponential function for modeling the electromechanical activation of the myometrium provides, according to the results obtained on our database, more accurate IUP estimates with respect to the methods previously proposed; the estimate accuracy is assessed by the correlation coefficient and the RMSE with the IUPC signal. On average, the logarithmic model, that does not account for an asymptotic behavior of the IUP, provides more accurate estimates than the exponential model.

Our study preliminarily suggests that the measure uncertainty of the described noninvasive IUP estimation methods can be comparable to the uncertainty of IUPC invasive measurements discussed in the literature. However, future work will be dedicated to the extensive clinical studies that are required for assessing the possibility of introducing noninvasive IUP recordings based on EHG signal analysis in everyday clinical practice.

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